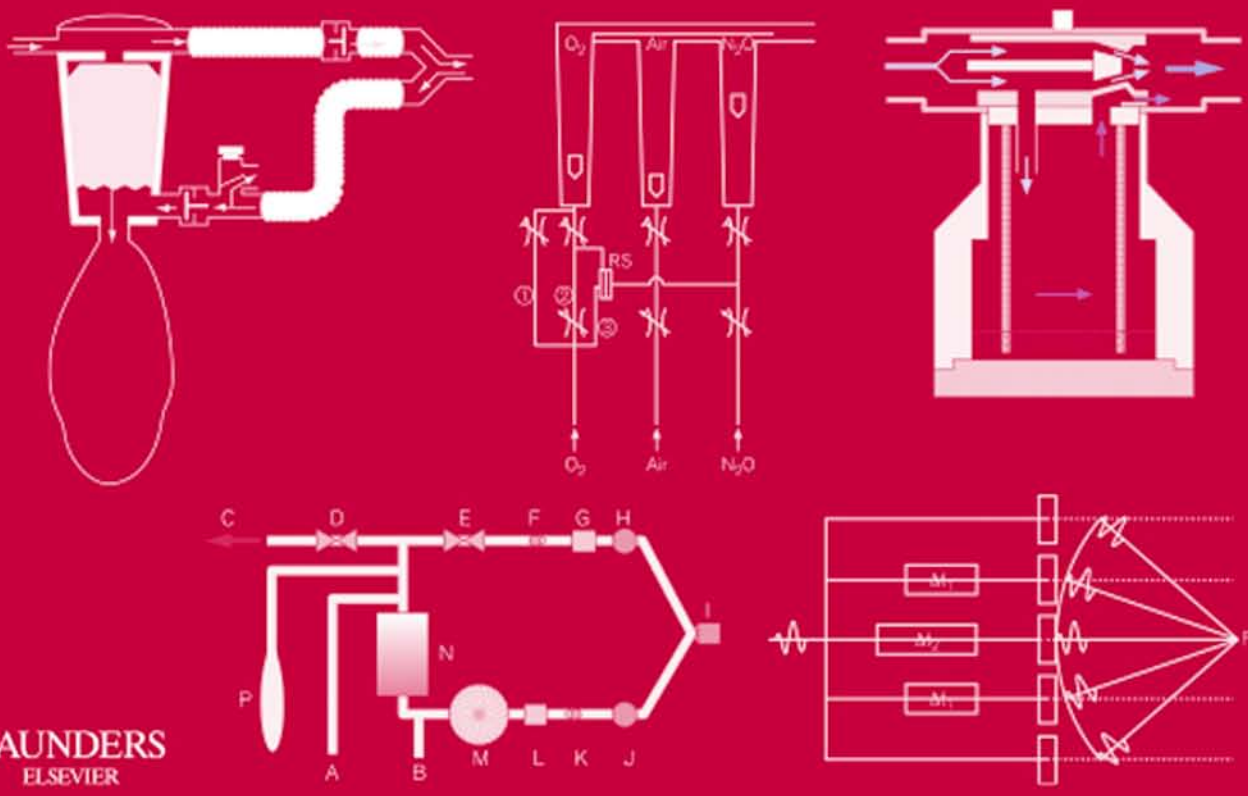
Andrew J Davey and Ali Diba

WARD'S

SIXTH EDITION

ANAESTHETIC EQUIPMENT



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ANAESTHETIC
EQUIPMENT

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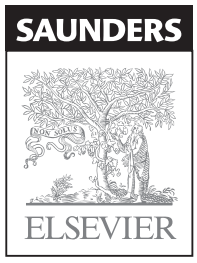
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Preface
to the Sixth Edition

The sixth edition of this book has built on the successful format of the previous edition. The latter was a departure from a dual authorship to a collection of contributions from numerous individuals with a specialist interest in order to cover all the chosen topics in depth and with up-to-date information. Advances in printing have allowed the use of colour photos and illustrations to enhance the text in a way that would have been prohibitively expensive in previous editions. The use of the Internet, electronic mail, portable document format (pdf) submissions for editing and proof reading have revolutionized data gathering for the book. Video conferencing between the editors has saved the world's carbon footprint by reducing commuting for face-to-face meetings! As always, the manufacturers have been magnificent in providing technical help and illustrative material. Our gratitude goes also to the editorial and production staff at Elsevier who have been immensely supportive and patient with the editors, recognizing that both have full time jobs in anaesthesia as well as their editorial responsibilities.

This book has two main purposes. The first is to provide a simple, yet comprehensive, explanation of the function of items of anaesthetic equipment to ensure their safe use in clinical practice by anaesthetists. The second is to provide a source of reference for trainee anaesthetists that covers the relevant syllabus required for the Primary and Final Fellowship Examinations in anaesthesia in the UK and Ireland. The book, hopefully, will also be of interest to anaesthetic assistants, electronic and biomedical engineers in hospitals, manufacturers' representatives and those involved in anaesthesia in other countries where similar equipment is used.

Many chapters have been extensively revised to include new developments and to eliminate obsolete items. New authors have been invited to build on the material provided by previous contributors. We hope they accept our additions, deletions and editorial incursions into their chapters, all of which we felt would benefit the tone and balance of the book as a whole.

Previous editions contained introductory chapters on basic physics relevant to equipment. After much soul searching, these have been dropped as they are covered more comprehensively by specific books within the Elsevier catalogue.

There has been an occasional deliberate repetition of material where the content of some chapters has overlapped. This caters for a reader who may wish to peruse individual chapters rather than read the book from cover to cover.

Finally, the editors have continued to try to follow the style, syntax and grammar of the earlier editions overseen by the late Dr. Crispian Ward whose brainchild this book is. We hope he would be pleased.

Andrew J Davey and Ali Diba

Acknowledgments

We are indebted to the many manufacturing companies who have provided us with essential data, illustrations, photographic material and loaned equipment. Some of the illustrations have, with permission, been altered to ensure that they conform to a uniform style. We are also grateful to the many personnel of those manufacturing companies, ranging from senior technical staff to representatives and service engineers, who tirelessly answered our repeated queries and requests. Also, a particular thanks must go to the Medical Photography Department of the Queen Victoria Hospital, East Grinstead for their forbearance in graciously accommodating endless requests for photography.

Dedication

This book is dedicated to the memory of Crispian Ward, late Consultant Anaesthetist, Huddersfield Royal Infirmary, who sadly died in 2002. In the early 1970s, he was one of the first anaesthetists to realize that an understanding and working knowledge of anaesthetic equipment was essential in order to provide the highest standards of safety in anaesthesia. He, therefore, set out to write a suitable book on the subject (the first of its kind in the UK) and applied his many talents to this work. He was a genius with the written word. He had the ability to explain and reveal in short readable sentences. Added to this were his remarkable powers of concentration, enthusiasm and meticulous attention to detail. Those copy editors employed by the publishers to read through his work were made virtually redundant, as he supplied almost flawless manuscripts. He was the scourge of typesetters by insisting that text and illustrations should appear side by side even if this entailed a major revision of a chapter's layout. The first two editions were written exclusively by him whilst working as a consultant in a busy district general hospital. This was a formidable achievement.

As the subject matter expanded and Cris reached retirement, he enlisted the help of two younger anaesthetic colleagues – John Moyle (also an electrical engineer) and myself – for the next two editions. Although he retained editorial command, he did this with such great diplomacy and tact that we were not slighted by any alterations that he requested. He was not in favour of books written by multiple contributors, as he felt this would affect the style of the written word! Sadly, the subject matter has expanded to such an extent that in order to provide the depth of knowledge required this is no longer possible. However, the present editors have attempted to follow his style and hope he would be pleased with the result. He is greatly missed.

Andrew J Davey

Abbreviations

AAGBI	Association of Anaesthetists of Great Britain and Ireland
AC	Alternating current
ACOP	Approved Code of Practice
ADC	Analogue to digital converter
AEP	Auditory evoked potential
AGSS	Anaesthetic gas scavenging system
AICD	Automatic implantable cardioverter-defibrillator
ANSI	American National Standards Institute
APL	Adjustable pressure-limiting (valve)
ARX	Autoregressive model with exogenous input
ASB	Assisted spontaneous breathing
ASTM	American Society for Testing and Materials
ATLS	Advanced trauma life support
atm	Atmosphere (unit of pressure)
AVSU	Area valved service unit
BBV	Blood-borne virus
BCGA	British Compressed Gas Association
BET	Bolus elimination and transfer
BiPAP	Bi-level positive airway pressure
BIS	Bispectral index
BP	Blood pressure
BP	British Pharmacopoeia
BPEG	British Pacing and Electrophysiology Group
BPM	Breaths per minute
BS	British Standard
BSI	British Standards Institute
BSP	British Standard Pipe (screw thread)
BSR	Burst suppression ratio
BTPS	Body temperature and pressure, saturated
C	Coulomb
cm H_2O	Centimetres of water (unit of pressure)
°C	Degrees Celsius
CBF	Cerebral blood flow
CCD	Charge coupled device
CE	Conformite Européene
CEN	Comité Européene de Normalization
CFCs	Chlorofluorocarbons
CFAM	Cerebral function analysing monitor
CGO	Common gas outlet
CLAN	Closed loop anaesthesia
CLS	Cryogenic Liquid Systems
CMRR	Common mode rejection ratio
CMV	Controlled minute ventilation
CNST	Clinical Negligence Scheme for Trusts
CO	Cardiac output
COELCB	Current-operated earth-leakage circuit breaker
COSHH	Control of Substances Hazardous to Health
CPAP	Continuous positive airway pressure
CPU	Central processing unit
CRT	Cathode ray tube
CSA	Compressed spectral array
CSSD	Central Sterile Supply Department
CVP	Central venous pressure
DAC	Digital to analogue converter
DC	Direct current
DFT	Defibrillation threshold
DLT	Double lumen tube
DoH	Department of Health (Replaces **DHSS** Department of Health and Social Security)
DISS	Diameter Indexed Safety System (USA)
DSA	Density spectral array
EBME	Electronic and Biomedical Engineering (Department)
ECG	Electrocardiogram
EEG	Electroencephalogram
EMG	Electromyogram
EMI	Electromagnetic interference
EMO	Epstein Macintosh Oxford
EN	Norme Européenne
EPROM	Eraseable programmable read only memory
EU	European Union
eV	Electron volt
FDA	Food and Drug Administration (USA)
FEV	Forced expiratory volume

FEV_1	Forced expiratory volume in 1 second
FFT	Fast Fourier Transformation
FG	French gauge
FGF	Fresh gas flow
FRC	Functional residual capacity
FT	Flow time
FTc	Corrected flow time
g	Gauge pressure (as opposed to absolute pressure)
GEB	Gum-elastic bougie
GHTF	Global Harmonization Task Force
HC	Health Canada
HEI	Health Equipment Information (issued by the DoH)
HFPPV	High-frequency positive pressure ventilation
HCMs	Hundreds of cubic metres
HMEF	Heat and moisture exchange filter
HME	Heat and moisture exchanger
ICD	Implantable cardioverter defibrillator
ICU/ITU	Intensive care unit/Intensive therapy unit
ID	Internal diameter
IEC	International Electrotechnical Commission
ILCOR	International Liaison Committee on Resuscitation
IMV	Intermittent mandatory ventilation
I/O	Input or output
IPPV	Intermittent positive pressure ventilation
IR	Infrared
ISO	International Organization for Standardization
K	Degrees Kelvin (always stated without degree symbol)
kg	Kilogram force
kPa	Kilopascal (European standard measurement for pressure)
LASER	Light Amplification by Stimulated Emission of Radiation
LCD	Liquid crystal display
LED	Light-emitting diode
LMA	Laryngeal mask airway
LOFT	Line Orientated Flight Training
MAP	Mean airway pressure
MDAPE	Median absolute performance error
MDPE	Median performance error
MF	Median frequency
MGPS	Medical gas pipeline services
MHRA	Medicines and Healthcare products Regulatory Agency
MHz	Megahertz
MLAEP	Mid-latency auditory evoked potential
mmHg	Millimetres of mercury (unit of pressure)
MMV	Mandatory minute volume
MORE	Manufacturers' On-line Reporting Environment
MRI	Magnetic Resonance Imaging
N	Newton
NAO	National Audit Office
NASPE	North American Society of Pacing and Electrophysiology
NEEP	Negative end-expiratory pressure
NELH	National Electronic Library of Health
NIBP	Non-invasive blood pressure (monitoring)
NICE	National Institute for Clinical Excellence
NiCd	Nickel-cadmium
NiMH	Nickel-metal hydride
NIOSH	National Institute for Occupational Safety and Health (USA)
NIST	Non-interchangeable screw threaded (connection)
NIPPV	Non-invasive positive pressure ventilation
NO	Nitric oxide
NPSA	National Patient Safety Agency
NRV	Non-rebreathing valve
O	Ohm
OD	Outside diameter
OIB	Oxford inflating bellows
OMV	Oxford Miniature Vaporizer
Pa	Pascal (unit of pressure)
PAC	Pulmonary artery flotation catheter
PAWP	Pulmonary artery wedge pressure
PC	Patient-controlled analgesia
PCEA	Patient-controlled epidural analgesa
PCV	Pressure controlled ventilation
PDPH	Post-dural puncture headache
PDT	Percutaneous dilatational tracheostomy
PEEP	Positive end-expiratory pressure
PEF	Peak expiratory flow
PIP	Peak inspiratory pressure
PNS	Peripheral nerve stimulator
PP	Pause pressure
ppm	Parts per million
PSA	Pressure Swing Absorber
psi	Pounds per square inch (US standard measurement for pressure)
PSV	Pressure support ventilation
PTFE	Polytetrafluorethylene (Teflon)
PV	Peak velocity
PVC	Polyvinyl chloride
RAM	Random access memory
RCCB	Residual current circuit breaker
RF	Radiofrequency
RMS	Root mean squared
ROM	Read only memory
RP (device)	Residual pressure device
RSA	Respiratory sinus arrhythmia
SABS	Safety Alert Broadcast System
SEF	Spectral edge frequency
SI (units)	Système International d'Unites (International System of Units)
SIB	Self-inflating bag

SIMV	Synchronized intermittent mandatory ventilation
SMA	Synthetic Medical Air
SSEP	Somatosensory evoked potential
SV	Stroke volume
SVP	Saturated vapour pressure
swg	Standard wire gauge
TAP	Transoesophageal atrial pacing
TCD	Transcranial Doppler ultrasonography
TCE	Temperature compensating element
TCI	Target controlled infusion
TFT	Thin film transistor
TIVA	Total intravenous anaesthesia
TOF	Train of four
Torr	(Torricelli) mmHg pressure (also used for subatmospheric pressures)
TSE	Transmissible Spongiform Encephalopathies
TSSU	Theatre Sterile Supply Unit
TT	Tracheal tube
TTJV	Transtracheal Jet Ventilation
TWA	Time-weighted average
USP	United States Pharmacopeia
UV	Ultraviolet
vCJD	Variant Creutzfeldt–Jakob disease
VF	Ventricular fibrillation
VIC	Vaporizer in circle
VIE	Vacuum insulated evaporator
Vmax	Maximal velocity
VOC	Vaporizer out of circle
WEL	Workplace exposure limit

Chapter 1

The supply of anaesthetic and other medical gasses

Hubert Bland, Carrie Borton and Syed Jafri

CHAPTER CONTENTS

Properties of medical gasses	**1**
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Under both the Medicines Act 1968 (UK) and more recent European legislation [EC 2001/83], medical gasses are classified as medicinal products. As such, they can only be produced by companies holding a manufacturer's licence and sold by agents holding a *marketing authorization* (previously known as a Product Licence). In the UK, these licences are issued by the Medicines and Healthcare products Regulatory Agency (MHRA). Similar regulatory bodies exist in most countries.

Medical gasses for use in anaesthesia and critical care are generally supplied to hospitals either in bulk (e.g. synthetic air, medical air and oxygen) or in individual cylinders collectively attached to manifolds (e.g. oxygen, nitrous oxide, medical air, Entonox and occasionally Heliox). In both cases the gasses are then delivered through a pipeline system to wall or pendant outlets. Where pipeline supplies are unavailable, for instance during patient transfers, the gasses are supplied directly from individual cylinders.

Oxygen can also be supplied by means of an oxygen concentrator, although this would be considered only where a bulk supply was either unavailable or impractical (e.g. due to geographical constraints).

Whilst the distribution of a pharmaceutical product in a highly pressurized metal cylinder, which is returned to the manufacturer for refilling, remains a unique concept in drug delivery, there have until recently been relatively few advances in the manufacturing, packaging, delivery and application of the products. The next decade will undoubtedly see the increasing clinical use of gasses such as Heliox, xenon and carbogen, but these novel agents are likely to generate new challenges – for example, the adoption of xenon, a potential neuroprotectant, in anaesthesia will bring with it problems associated with ventilating with a gas approximately five times denser than air.

PROPERTIES OF MEDICAL GASSES

The properties of some of the common medical gasses are summarized in Table 1.1. The newer gasses and their potential therapeutic indications are briefly discussed below.

Details of carbogen (5% CO_2 + 95% O_2), lung function gasses, medical gasses for laser surgery and other agents not listed here are available from the manufacturers. Extensive information on the properties of all gasses and data sheets containing up-to-date product information can be obtained from the BOC Medical Gasses website *www.bocmedical.com*.

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Table 1.1 Physical properties of common pressurized gasses

	OXYGEN	NITROUS OXIDE	CARBON DIOXIDE	XENON	NITRIC OXIDE	CARBON MONOXIDE	HELIUM
Physical state in cylinder	Gas	Liquid	Liquid	Gas	Gas	Gas	Gas
Molecular weight	32	44	44	131	30	28	4 (He)
Melting point (°C)	N/A	–90.81	–56.6	–112	–164	–205	N/A
Boiling point (°C)	–183	–88.5	–78.5*	–108	–152	–192	–269
Critical temperature (°C)	–118.4	36.4	30	16.6	–93	–140	–268
Relative density, gas (air = 1)	1.04	1.5	1.52	4.5	1	1	0.14
Relative density, liquid (water = 1)	N/A	1.2	0.82	N/A	1.3	N/A	N/A
Vapour pressure at 20°C (bar)	N/A	50.8	57.3	N/A	N/A	N/A	N/A
Solubility water (mg/mL)	N/A	2.2	2000	644	67	30	1.5
Appearance/colour	Colourless gas	Colourless gas	Colourless gas	Colourless gas	Colourless gas	Colourless gas	Colourless
Odour	None	Sweetish	None	None	Pungent	None	None
Other data		Gas/vapour heavier than air. May accumulate in confined spaces, particularly at or below ground level	Gas/vapour heavier than air. May accumulate in confined spaces, particularly at or below ground level	Gas/vapour heavier than air. May accumulate in confined spaces, particularly at or below ground level. Note that due its density, Xe will flow through standard flowmeters more slowly. A conversion factor of 0.468 should be applied	Currently supplied in N_2 at less than 1000 ppm	Currently only supplied at mixtures of less than 0.3% in air and helium	Use the appropriate conversion chart when using oxygen flowmeters

N/A, not applicable. *Sublimation.

Note: all values STP. Mixed gasses, e.g. Heliox, will have different physical properties. For exact values, please contact the manufacturer.

Heliox

Heliox is a generic term for any blend of oxygen and helium. Currently only a specific premixed blend – 21% oxygen and 79% helium known in the UK as Heliox21 – has approval for distribution. Heliox21 is licensed for the treatment of respiratory obstruction and, although traditionally confined to emergency use, is increasingly being used by other hospital departments and the emergency services.

Heliox21 has different physical properties compared to air (which is sometimes called nitrox for comparison) and these cause it to behave differently when used with anaesthetic and respiratory equipment. Most importantly:

- It has a lower density than air (0.42 kg/m^3 vs 1.22 kg/m^3 at 15°C and 1 atm) and hence a lower Reynolds number in any given situation, resulting in a higher likelihood of laminar gas flow characteristics.
- It has a higher thermal conductivity than air, which is of significance when its gas flow measurement is based on this principle (see Chapter 2, Hot wire anemometry).

If administered via standard air or oxygen flowmeters, flow readings will be inaccurate unless a conversion factor is applied. Appropriate conversion tables are available from the manufacturer. Flow/volume measurements on ventilators are similarly affected, hence only readings from machines designed for ventilation with helium mixtures can be assumed to be accurate.

Xenon

Xenon is not a standard medical product and is provided as a 'special' gas. It demonstrates anaesthetic and neuroprotective activity. However, since it is five times denser than air, mechanical ventilation is only possible with specially designed equipment. Furthermore, it is supplied at a lower pressure than oxygen and Heliox21, and, consequently, should be used with compatible low-pressure regulators. Its density will slow down its flow through narrow tubes, including standard flowmeters. A conversion factor of 0.468 should be applied. Until a licence is granted for its use in anaesthesia or other applications, usage in medicine is restricted to research applications.

Nitric oxide

Although toxic at high concentrations, when given in very low dilutions (8–50 ppm) nitric oxide acts as a selective pulmonary vasodilator and has been used extensively in the paediatric intensive care setting. Because of its acidic properties in the presence of moisture, administration is restricted to specialized, compatible equipment. Research is currently focusing on its use as an immuno-modulator, in platelet function alteration and for domiciliary use.

Carbon monoxide

Carbon monoxide is being investigated in connection with a number of therapeutic areas (suppression of inflammation, vasodilatation, cytoprotection and organ transplantation). Chromium-plated equipment should never be used with CO because of the risk of generating highly toxic chromium containing compounds such as chrome carbonyl.

MEDICAL GAS CYLINDERS

Modern cylinders are manufactured using lightweight strong chrome-molybdenum steels which, as well as conforming to stringent material standards, can be filled to pressures of up to 300 bar g (where (g) denotes gauge pressure; see Chapter 2). Cylinders were previously made of the much heavier low-carbon steels: few of these are in circulation any longer in the UK.

All-steel cylinders are perfectly adequate for applications where weight is not an issue; such as on cylinder manifolds. In situations where portability is important, lighter weight composite (hoop wrap) cylinders are used. These are constructed from two or more different materials; commonly either lightweight steel or, more commonly, an aluminium liner which is then strengthened by wrapping a filament material, such as Kevlar or carbon-fibre, coated in epoxy resin circumferentially along the parallel length of the cylinder. The cylinders combine enormous strength, with low weight and can be filled to pressures up to 300 bar g.

In some locations, for example in rooms containing MRI scanners, it is not possible to use cylinders containing steel as these are ferromagnetic and can be uncontrollably accelerated into the MRI with potentially fatal consequences. Aluminium cylinders with special non-ferromagnetic pin index valves are available for this application, but it is recommended that, where possible, gasses should be piped into the unit from outside.

Cylinder sizes

Technically, cylinders are defined by their water capacity and range between 1.2 L and 47.2 L, and are identified by a size code ranging from C to J (Fig. 1.1). This notation incorporates a pressure aspect to give an indication of available gas volume. Tables 1.2–1.7 give details for oxygen, nitrous oxide, Entonox, carbon dioxide, Heliox21, xenon, nitric oxide and carbon monoxide cylinders.

Cylinder filling and maintenance

Most gasses such as oxygen, medical air, helium and Heliox21 are stored in cylinders in a compressed gaseous

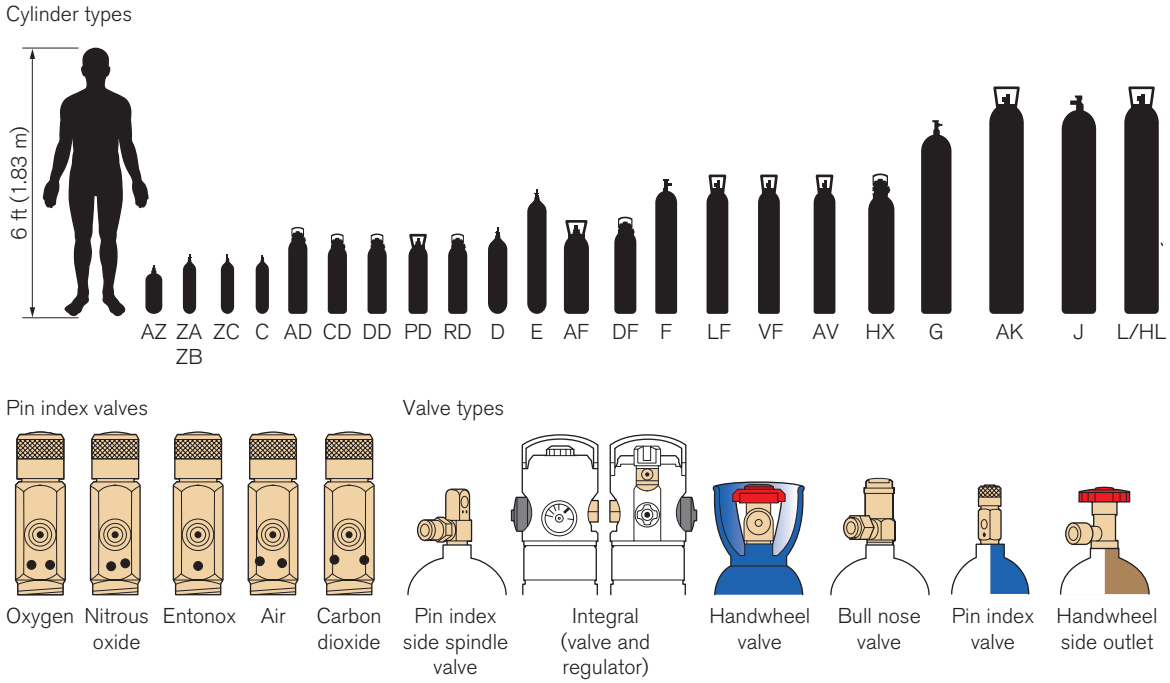


Figure 1.1 Cylinder types and sizes, valves types and pin index valves. *(from BOC Medical, with permission).*

state and are normally filled by pressure. Nitrous oxide and carbon dioxide, however, are liquefied gasses under pressure. The liquid is in equilibrium with the gas, the pressure being dependent on the temperature. These gasses are charged by weight and not by pressure. The maximum *weight* of gas filled into the cylinder divided by the *weight* of water that would completely fill the cylinder is called the maximum filling ratio and is governed by legislation. This 'fill ratio', termed 'fill density' elsewhere, is critical. In the UK this value is 0.75 at 15.5 °C for nitrous oxide and carbon dioxide. Note that due to the difference in the densities of liquefied gasses and water this ratio is not the same as the proportion of the volume of the cylinder filled by the liquid phase, which is nearer 90 to 95% for nitrous oxide. If there is insufficient gas space left in the cylinder after filling, a comparatively small increase in temperature will cause a significant increase in pressure and could, in extreme circumstances, cause the cylinder to rupture.

Cylinders of nitrous oxide and carbon dioxide should always be used in the vertical position with the outlet uppermost; this prevents liquefied gas from being vented and causing cold thermal burns or equipment damage. Also when a high flow rate is drawn, the temperature of the liquefied gas will fall – this can result in a significant pressure drop which may cause poor performance when used with equipment such as a cryogenic probe. A good indicator of this phenomenon is the appearance of water vapour condensing/freezing on the outer surface of the cylinder. The effect is most common on smaller liquefied gas cylinders up to size F (9.43 litres).

Some applications require carbon dioxide in liquid form. This is supplied using cylinders fitted with a 'dip tube' connected to the cylinder valve, which allows liquid from the bottom of the cylinder to be drawn up through the valve (Fig. 1.2). These cylinders, supplied in F size and known as LF, have a white stripe down the length of the cylinder body. For other CO_2 applications where vapour is required, VF cylinders should be used. It is important to ensure that the correct cylinder is used to prevent damage to equipment.

Checking for cylinder contents is done in one of two ways: compressed gasses such as oxygen, medical air, helium, Heliox and Entonox (which, although 50% nitrous oxide, remains as a gas in the cylinder under normal ambient temperatures) are assessed using a pressure gauge, as the pressure in the cylinder is directly proportional to the volume. By contrast, the contents of cylinders containing liquefied gasses (pure nitrous oxide and carbon dioxide) can only be determined by weight, as the pressure only begins to fall once all the liquid is exhausted. Subtracting the *tare* weight (stamped on the neck of the cylinder) from the total weight will give an estimate of contents.

Table 1.2 Relative sizes and specifications of commonly used oxygen cylinders

CYLINDER SIZE	C	CD/DD	D	E	F	HX	G	J	ZX	ZA
Contents (L)	**170**	**460**	**340**	**680**	**1360**	**2300**	**3400**	**6800**	**3970**	**300**
Nominal cylinder pressure at 15°C (bar)	137	230	137	137	137	230	137	137	300	300
Valve type	Pin index	Integral	Pin index	Pin index	Bull nose	Integral	Bull nose	Side spindle pin index	Integral	Integral
Water capacity (litres)	1.2	2	2.32	4.68	9.43	10	23.6	47.2	10	1
Dimensions (mm)	430 × 189	520 × 100	535 × 102	865 × 102	930 × 140	940 × 140	1320 × 178	1520 × 229	940 × 143	366 × 85
Empty weight (kg)	2.0	3.0	3.4	5.4	14.5	15	34.5	68.9	10	1.2

Table 1.3 Relative sizes and specifications of commonly used nitrous oxide cylinders

CYLINDER SIZE	C	D	E	F	G	J
Contents (L)	**450**	**900**	**1800**	**3600**	**9000**	**18000**
Nominal cylinder pressure at 15°C (bar)	44	44	44	44	44	44
Valve type	Pin index	Pin index	Pin index	Handwheel 11/16″ × 20 tpi	Handwheel 11/16″ × 20 tpi	Handwheel 11/16″ × 20 tpi
Dimensions (mm)	430 × 189	535 × 102	865 × 102	930 × 140	1320 × 178	1520 × 229
Empty weight (kg)	2.0	3.4	5.4	14.5	34.5	68.9

Table 1.4 Relative sizes and specifications of commonly used Entonox cylinders

CYLINDER SIZE	D	CD	ED	F	HX	G
Contents (L)	**500**	**440**	**700**	**2000**	**2200**	**5000**
Nominal cylinder pressure at 15°C (bar)	137	137	217	137	137	137
Valve type	Pin index	Integral	Integral	Side spindle pin index	Integral	Side spindle pin index
Dimensions (mm)	535 × 102	520 × 100	520 × 100	930 × 140	940 × 140	1320 × 178
Empty weight (kg)	3.4	2.7	2.8	14.5	15.5	34.5

Entonox, a mixture of 50% oxygen and 50% nitrous oxide, exists as a gas. The pseudo critical temperature of Entonox in pipelines at 4.1 bar is below –30°C. Nitrous oxide in an Entonox cylinder, however, begins to separate out from Entonox if the temperature falls below –6°C. A homogenous mixture is again obtained when the temperature is raised above 10°C and the cylinder is agitated.

Table 1.5 Relative sizes and specifications of commonly used carbon dioxide cylinders

CYLINDER SIZE	C	E	VF	LF
Contents (L)	**450**	**1800**	**3600**	**3600**
Nominal cylinder pressure at 15°C (bar)	50	50	50	50
Valve type	Pin index	Pin index	Handwheel 0.86″ × 14 tpi	Handwheel 0.86″ × 14 tpi
Dimensions (mm)	430 × 89	865 × 102	930 × 140	930 × 140
Empty weight (kg)	2.0	5.4	14.5	14.5

Table 1.6 Relative sizes and specifications of commonly used Heliox21 cylinders

CYLINDER SIZE	HL	HX
Contents (L)	**8200**	**1780**
Nominal cylinder pressure at 15°C (bar)	200	200
Valve type	Side outlet	Integral
Dimensions (mm)	1540 × 230	940 × 140
Empty weight (kg)	50	15.5

Cylinder identification

The correct method of identifying the contents of a cylinder is to read the collar identification label (rather than assuming that the colour of the cylinder or the type of valve fitted reliably indicates gas inside it). The label (Fig. 1.3) is a legal requirement and contains all the key information for the user:

- product name, chemical symbol and pharmaceutical form of the product
- product specification
- hazard warning diamond(s)
- product licence number
- cylinder contents in litres
- maximum cylinder pressure
- cylinder size code
- directions for use and information for storage and handling.

The cylinder label also has a unique batch label (Fig. 1.4), which contains: the batch number; fill and expiry date; and the size and type of gas. The label is changed every time the cylinder is filled and serves two important functions: (a) it provides vital information for a batch recall should the cylinder be involved in an incident; and (b) it provides information for proper cylinder rotation.

Cylinder testing

All cylinders must undergo hydraulic testing and internal inspection at regular intervals, to ensure they remain safe to use. The test is carried out every 10 years for steel

Table 1.7 Relative sizes and specifications of commonly used xenon, nitric oxide and carbon monoxide cylinders

	XENON	NITRIC OXIDE	CARBON MONOXIDE
Cylinder size	No medical standard size currently available	Various – depending on manufacturer	No standard cylinder size currently available for medical use – provided as a special gas
Nominal cylinder pressure at 15°C (bar)	Filled to low pressures of around 30 bar	Filled to a lower pressure than standard medical gas cylinders (around 30 bar)	Relatively low pressure fill – currently available between 8 and 36 bar
Valve type	Low pressure regulator required	Special equipment needed	Note: Will require low pressure compatible regulators

cylinders and every 5 years for composite cylinders. A colour-coded plastic ring between the valve and the cylinder neck indicates when the next test date is due. In general, cylinders have a long service life and tend to be withdrawn for reasons of technical obsolescence rather than deterioration per se.

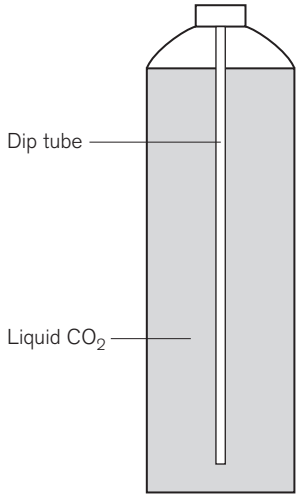


Figure 1.2 LF type CO_2 cylinder with dip tube for delivering liquid CO_2.



Figure 1.3 Cylinder label.

Colour coding

Medical cylinders in the UK conform to colour codes specified in ISO 32:1972 and EN 1089-3. The colour coding relates only to the shoulder of the cylinder. Figure 1.5 shows the colour codes for medical gas cylinders in the UK.

Cylinder valves

Medical cylinder valves can generally be categorized as follows:

- Non-integral valves (which require manual attachment of an external pressure regulator):
  - Small pin index and side spindle pin index outlet valves
  - Bull nose outlet valves
  - Handwheel actuated valves
- Integral valves (which have a built in regulator and possibly flowmeter).

Pin index system

Small pin index valves (Fig. 1.6) are fitted to small cylinders (less than 5 litre capacity) which are commonly connected directly to medical equipment such as anaesthetic machines. Newer designs using a thumbwheel do away with the need for a spanner to operate the valve.



Figure 1.4 Batch label.

Gas	Identification markings on cylinder shoulder		BOC product information leaflets†
Oxygen		White	MED/004041
Nitrous oxide		Blue	MED/004040
Entonox (50% N_2O/50% O_2)		Blue/white	MED/004042
Air		Black/white	MED/004038
Oxygen/carbon dioxide mixture (95% O_2/5% CO_2)		Grey/white	MED/004035
Helium/oxygen mixture (79% He/21% O_2)		Brown/white	MED/004034
Carbon dioxide		Grey	MED/004039
Helium		Brown	MED/004037

* Note: Cylinder identification colours are those specified in ISO 32 (1977) and BS EN 1089-3 : 2004
† Up-to-date product information leaflets can be downloaded from http://www.bochealthcare.co.uk/en/safety/sds/index.shtml

Figure 1.5 Cylinder colour codes.

Side spindle pin index valves (Fig. 1.6) These are fitted to large cylinders of medical oxygen, medical air, Entonox used in pipeline manifolds and F size Entonox cylinders.

Both types of pin index valves conform to BS EN ISO 407:2004 and adopt an indexed outlet system which incorporates a gas-specific combination of holes positioned to correspond to pins located on the receiving equipment, making it impossible to connect the cylinder to an incorrect gas connection. Fig. 1.1 shows the different pin positions. The pin index system also prevents charging with the wrong gas, as the gas suppliers use the same non-interconnectable system for their filling connections.

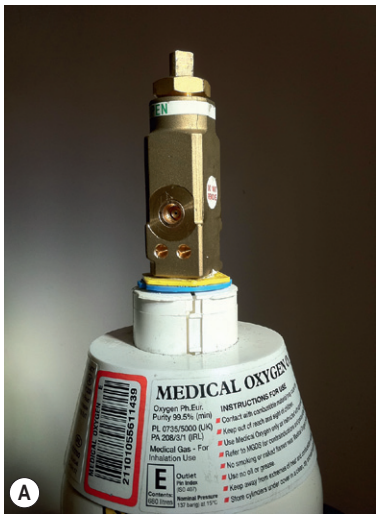
Pin index cylinders require a washer (seal) between the face of the cylinder valve outlet and the equipment to which it is fitted. This bonded non-combustible seal, known as a 'Bodok' washer (Figs 1.7 and 1.8), must be kept clean and should never become contaminated with oil or grease. If a gas tight seal cannot be achieved by moderate tightening of the screw clamp, it is recommended that the seal be renewed. Excessive force should never be used.

Bull nose outlet valve

This type of valve (Fig. 1.9) is fitted to F and G size cylinders including medical oxygen, medical air, helium, and mixed gasses such as carbogen and Heliox. The valve is spindle key operated and has a 5/8-inch female outlet thread into which a regulator is fitted. The spindle mechanism is assembled in two parts. This permits a gas tight seal to be achieved without the use of excessive force and increases its operational life.

All bull nose valves are fitted with an RP (residual pressure) device, to ensure that a positive pressure of approximately 3 bar is retained in the cylinder (Fig. 1.10). This prevents the ingress of moisture should the valve be left open when the cylinder is empty. When connecting regulator equipment to the valve, the user should always adopt the proper connecting procedure:

- Ensure the 'O' ring (Fig. 1.11) fitted to the regulator is in good condition and hand tighten the regulator only.
- Once fitted, open the spindle valve slowly (to prevent a gas surge) at least one full turn and note the position of the regulator gauge needle.
- A simple test for leaks can be carried out by closing the cylinder spindle valve and noting any drop in pressure shown on the gauge. If a leak does exist, spraying the joints with a leak detection spray will identify it. If a leak is evident at the valve outlet, replace the 'O' ring and repeat the procedure. If the leak persists, fit a replacement regulator. (Note: It is vitally important that any leak detection spray has been approved by the cylinder supplier as being compatible with their equipment and the gas.)
- When changing an empty cylinder, close the cylinder valve and vent the regulator completely before attempting to disconnect it.



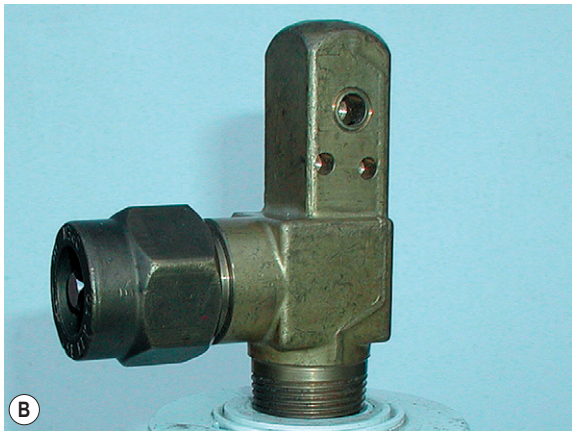


Figure 1.6 **A,** Small pin index valve on E size cylinder. **B,** Side spindle pin index valve.

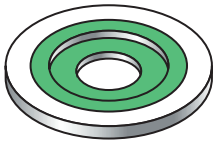


Figure 1.7 Bodok washer.

Handwheel valves

Large nitrous oxide cylinders for use on cylinder manifolds and carbon dioxide cylinders of size F and G are fitted with handwheel valves which are surrounded by a protective guard (Fig. 1.12), and have a gas specific, male thread, side outlet.

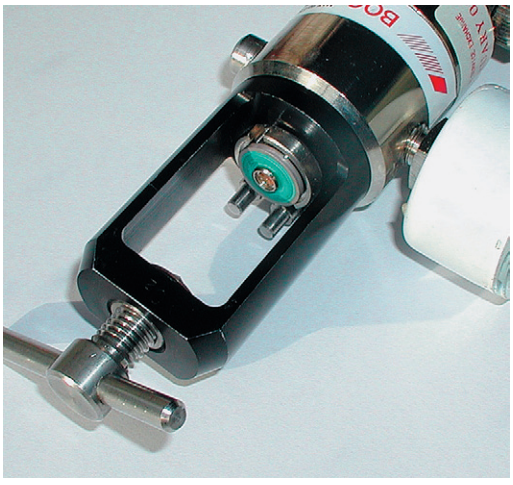


Figure 1.8 Bodok washer fitted to a regulator.



Figure 1.9 Bull nose cylinder valve.

Integral valves

The introduction of integral valves (Fig. 1.13), which have their own built-in regulator, has revolutionized the industry by greatly improving safety and eliminating the need for regulator maintenance by hospitals. They have a number of advantages:

- The regulator assembly is manufactured in a clean environment and is much less prone to particulate contamination.
- The built-in regulator eliminates the risk of incorrect regulator attachment which could result in damage to lower-pressure rated equipment.

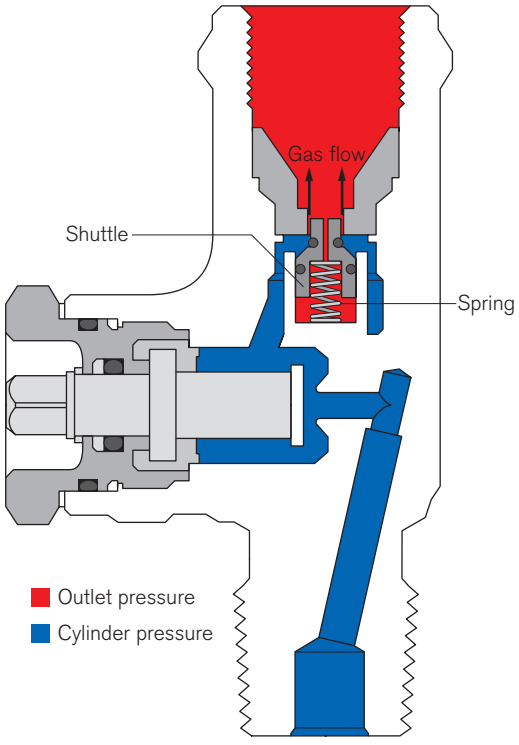


Figure 1.10 Schematic showing 'residual pressure' device of a bull nose cylinder valve.
Redrawn using material kindly provided by Müller Gas Equipment A/S, Denmark.



Figure 1.11 'O' ring fitted to regulator.



Figure 1.12 CO_2 cylinder with handwheel valve.



Figure 1.13 An integral valve without its guard.

- The valves are surrounded by guards which reduce the risk of damage and ease manual handling of the cylinder.
- They are considerably easier to use and have live pressure gauges built in for checking cylinder contents.
- The valves possess two outlets – a variable flow outlet (via a fir tree connector) and a fixed pressure outlet to connect equipment requiring a static driving pressure.

A commonly encountered version of the integral valve has a combination of a clickstop flowmeter giving 0–15 l min^{-1} in stepped increments and a female BS Schraeder connection at 4 bar g to provide high pressure gas. This type of valve is currently fitted to cylinders of up to 10 l water capacity. Care must be taken to ensure that the flowmeter selector dial is not 'parked' between click stops as this will result in a cessation of gas flow.

Material compatibility

Special care is needed in the selection of metallic and non-metallic materials used in the manufacture of cylinders as substances burn far more readily in an oxygen-rich environment. For example, the elastomers used in O rings, seats and seals must be resistant to burning in 100% oxygen and, on eventual combustion, must not generate toxic gasses. BS EN 15001 provides advice on materials which can be used safely.

Tamper evident seals

All cylinder valves are fitted with tamper evident seals when delivered. The seals are usually shrink wrapped around the valve or, in the case of integral valves, are in the form of a tear off. They identify cylinders as being full and should only be removed at the point of use. Bull nose valves have a protective cap, which should be replaced after use.

Storage of medical gas cylinders

Cylinders should be stored in accordance with recommendations detailed in HTM 02-01. They should not be stored with non-medical cylinders and the store should:

- be under cover (preferably inside) and not subjected to extremes of heat
- be kept dry, clean and well ventilated
- have good access for deliveries and a reasonably level floor surface
- allow segregation of 'full' and 'empty' cylinders
- permit separation of different gasses and cylinder sizes
- allow for strict stock rotation to enable cylinders with the oldest fill date to be used first
- be sited away from storage areas containing combustible materials or sources of heat or ignition
- have warning notices clearly posted prohibiting smoking or naked lights in the vicinity
- allow for large cylinders to be stored vertically in concrete pens and small cylinders to be stored horizontally in wooden or plastic racks to prevent damage to the cylinders
- not allow the temperature to fall below 10 °C where full Entonox cylinders are stored
- be designed to prevent unauthorized entry.

CYLINDER MANIFOLDS

Although few hospitals rely on cylinder manifolds for their main oxygen or medical air supply, they are still used in reserve systems and as the main source of nitrous oxide and Entonox supply. Whilst there are minor differences for each gas, in general the systems are designed and operate along the same principles.

The typical configuration consists of two equal banks of gas cylinders (one demarcated duty and one stand-by). These are arranged around a central control panel and provide a nominal output pressure of 4 bar (7 bar for surgical air). The change over from the 'duty' to 'stand-by' banks is normally automatic. The installation should also contain a manually operated reserve of at least two cylinders (Fig. 1.14) also stored in the manifold room. Any additional cylinders should be held in the general medical gas store.

The total storage capacity of the manifold should be equivalent to 1 week's supply; a minimum of 2 days' supply on each bank and 3 days' supply held in the reserve cylinders.

Table 1.8 gives nominal and usable capacities of cylinders commonly used on manifolds.

Cylinders are attached to the manifold via a copper tailpipe with a gas specific connection and seal. Each connection has a non-return valve fitted to enable single cylinders to be changed in the event of a leak or tailpipe rupture. The cylinders are secured by individual chains to a back bar. All cylinders on both the duty and stand-by banks should be fully open. The central control panel determines which of the banks is the active duty bank. When this bank falls to a pressure of 8 bar, it switches to the stand-by bank and indicates on the alarm panel that the duty bank is empty and the stand-by is running. The responsible person should then change the empty bank of cylinders.

If the empty bank is not changed or a manifold fault occurs, once the stand-by bank (now duty bank) pressure falls to 8 bar, a pipeline pressure fault will register. The main manifold should then be isolated via a shut-off valve and the emergency bank manually opened, until normal conditions can be restored. The following alarm states occur:

- a green 'normal' condition
- a yellow 'duty bank empty, stand-by running' condition
- a yellow 'duty bank empty, stand-by low' condition
- a yellow 'emergency/reserve banks low' condition
- a red 'pipeline pressure fault' condition.

Whilst it is common throughout mainland Europe for nitrous oxide to be supplied in bulk, in the UK the gas is still supplied via manifolds which have heaters fitted to the supply line to prevent freezing during periods of high demand.



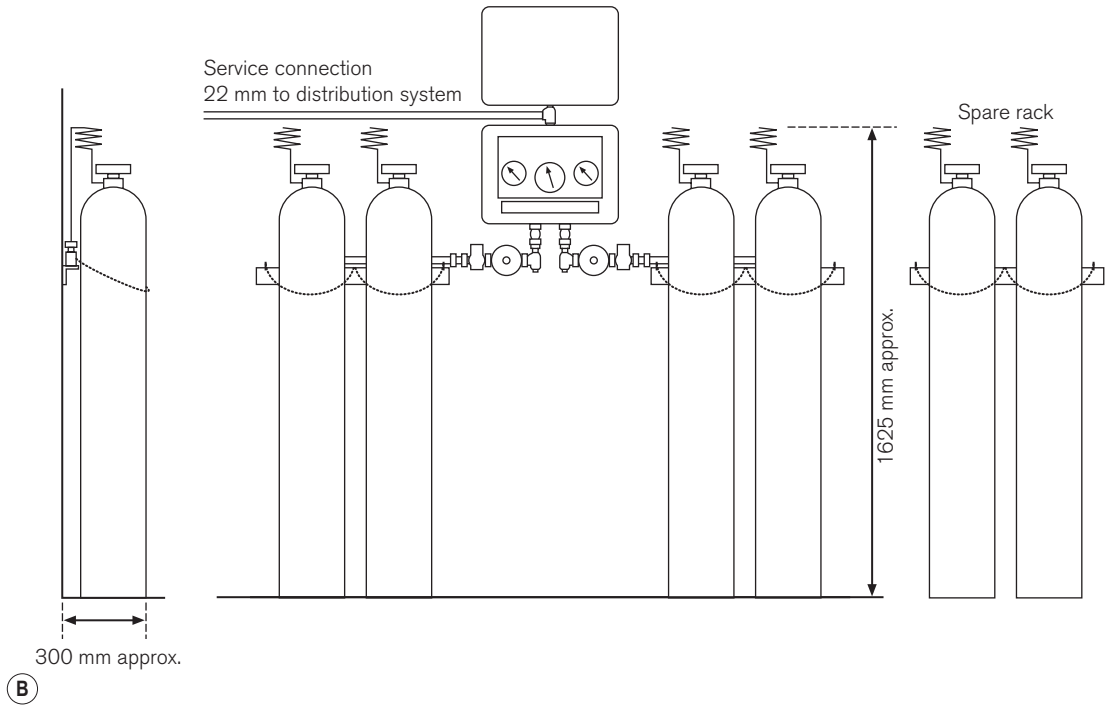


Figure 1.14 **A.** A nitrous oxide cylinder manifold. **B.** Schematic.

Table 1.8 Cylinder capacities

GAS	CYLINDER SIZE	NOMINAL CAPACITY	USABLE CAPACITY*
Oxygen	J	6800	6540
Nitrous oxide	G	9000	8900
Entonox	G	5000	4740
Medical air at 400 kpa	J	6400	6220
Medical air at 700 kpa	J	6400	5540†

*The usable figures are based on a residual pressure of 7 bar in the cylinders (†15 bar residual pressure).

Safety precautions

The manifold room should:

- be constructed from a fireproof material – either brick or concrete
- have ventilation at the top and bottom to permit the circulation of air
- ideally be located to enable a delivery vehicle access to prevent manhandling cylinders long distances
- be well lit
- be temperature controlled between 10 and 40°C. This is especially important in the case of an Entonox manifold as nitrous oxide and oxygen can separate out at temperatures below −6°C. It is advisable to let the cylinders stand for 24 hours in a temperature of more than 10°C before fitting them to the manifold
- only contain cylinders for use on the pipeline(s) located in the manifold room
- not be used as a general store
- have sufficient warning signs on the outside and inside of the building.

Only suitably trained persons should be permitted to change cylinders and an activity log should be completed when cylinders are changed.

BULK OXYGEN SUPPLY SYSTEMS

The first bulk medical oxygen systems were installed in the UK in the mid-1960s and have steadily increased in popularity since. Today almost all large hospitals have their piped medical oxygen supplied from an on-site, bulk oxygen supply facility. This reflects a steady rise in hospital oxygen consumption driven not only by the extension of piped oxygen from operating theatres and ICUs to most departments and wards, but also by significant changes in postoperative and ventilatory management. Currently an average 800-bed teaching hospital consumes around 500 million L of oxygen per year. It is difficult to relate to such large numbers, which is why supply companies refer to volume in hundreds of cubic metres (HCMs). Using the above example, the annual usage would be 5000 HCM or approximately 73500 J sized cylinders. Bulk oxygen systems are preferred owing to their ability to reliably deliver these volumes:

- at a lower overall gas cost
- without the labour-intensive distribution associated with compressed gas cylinders
- with greater on site storage capacity (at 15°C, one volume of liquid oxygen gives 842 times its volume compared with a manifold cylinder which delivers only 137 times its volume)
- with greater security of supply.

Whilst the volumes appear to be enormous, the actual cost per litre – around 0.0008 pence per litre in the UK – is extremely low. Unfortunately, increasing demand has necessitated the use of progressively larger bulk storage vessels, which can be difficult to site, especially in inner-city hospitals where space is at a premium and safety concerns may be difficult to mitigate.

Since the 1960s, the recommended 'on-site' storage capacity has increased from 6 to 14 days and is now based on a risk assessment which is referred to below.

Cryogenic liquid system (CLS)

The CLS system consists of:

- an insulated cryogenic storage vessel to store the bulk liquid oxygen
- an ambient heated vaporizer to convert the cryogenic liquid oxygen into a gas for supply to patients via a pipeline distribution system
- a control panel to control the pressure and flow of gas to the pipeline
- a telemetry system (see below).

The CLS can comprise:

- a single vessel containing operational stock with a cylinder manifold containing the secondary supply. Fig. 1.15 shows a schematic of a single-vessel installation

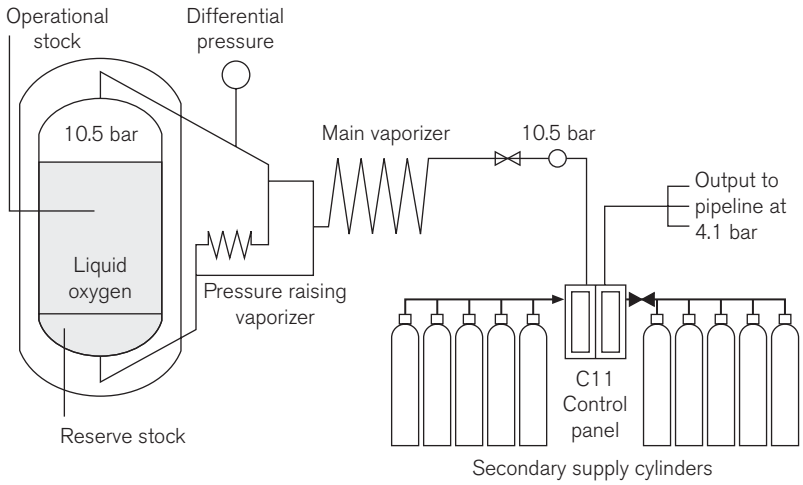


Figure 1.15 Simplified schematic of a single vessel cryogenic liquid system.



Figure 1.16 A twin vessel cryogenic liquid system installation.

- a main vessel containing operational stock with a second vessel either alongside or remotely located containing reserve stock, and a cylinder manifold containing emergency stock (Fig. 1.16).

A CLS should comply with recommendations in Chapter 6 of HTM 02-01 (part A) (Medical Gas Pipeline Systems) and take account of the criteria laid down in the European Standard BS EN ISO 7396, the British Compressed Gas Association (BCGA) Code of Practice CP 36 and the relevant UK legislation. HTM 02-01 also contains more detailed schematics of all types of CLS installations.

The basic function of a liquid oxygen vessel is to store cryogenic oxygen at −183°C, in what is effectively a vacuum flask; the inner vessel being made from stainless steel and the outer from carbon steel. Between the two vessels is a vacuum and the space is filled with a high-performance insulating material. The vessel and its associated controls are commonly known as a VIE (vacuum insulated evaporator).

Liquid oxygen sits in the bottom of the vessel whilst the gaseous oxygen floats above it at a pressure of 12–20 bar. Because it is impossible to maintain perfect insulation, the inner container is continually trying to draw heat from the atmosphere, though this is partially offset by the evaporation of liquid during use. If there is no demand, the pressure inside the vessel will rise, causing the safety relief valve to vent gas to atmosphere; to avoid this, the flow valves are designed to open under high pressure and permit gas to pass into the pipeline distribution line. Conversely, if demand is high, the pressure in the vessel will tend to fall. When this happens, liquid is withdrawn from the inferiorly located liquid valve and passed through a pressure-raising coil which raises the pressure to 10.5 bar.

During normal operation the liquid converts to a gas as it passes through a process vaporizer. This can be either a simple ambient vaporizer or duplex timed automatic switching vaporizers designed to allow one to operate whilst the second one defrosts. The C11 control panel (see Fig. 1.15) has duplicate regulators for security. These are designed to control the pressure at 4.1 bar for the main supply and 3.7 bar for the emergency cylinder supply. The control panel is designed to enable flows of up to 5000 L per minute from the main VIE supply and 1500 L per minute through the emergency cylinder manifold.

The control panel relays alarm conditions to a central alarm panel, usually located in the hospital telephone switchboard or other 24-h manned location, with duplicate alarm panels located in high acuity areas throughout the hospital, e.g. theatres, ITU and SCBU. The conditions can vary, depending upon the type of installation. A simple VIE with a cylinder manifold emergency would give the alarm conditions in Table 1.9.

Table 1.9 Alarm conditions for a simple VIE with a cylinder manifold backup (see text and Fig. 1.15)

STATUS/FAULT CONDITION	INDICATION	LEGEND
Normal operation	Green	Normal
Primary supply system operational stock empty Primary supply system reserve stock in use	Yellow	Liquid low Re-fill liquid
Primary supply system reserve stock empty Secondary supply system in use	Yellow	Re-fill liquid immediately
Secondary supply system low Lead secondary supply system content below 50%	Yellow	Change cylinders
Pipeline pressure fault (high, low)	Red	High pressure Low pressure

The VIE has a contents gauge which operates on differential pressure. The mass of the dependent liquid oxygen causes the pressure at the bottom of the vessel to be greater than that at the top and the gauge measures this difference and converts it into an analogue readout.

It is advisable to install a telemetry system to the CLS to provide continuous condition monitoring for both the supplier and the hospital CLS management.

Siting requirements

The installation should be located inside a fenced compound, be accessible to road tankers and be sited in accordance with the British Compressed Gasses Association (BCGA) code of practice and safety distance data provided by the installer. In general, all hazardous buildings, flammable materials, public access, vehicles and surface water drains, must be at least 5 m (and in the case of larger installations, 8 m) from the nearest point of the CLS compound. The compound floor and the hard standing area directly in front of the fill connection must be concrete and should be designed to contain any liquid spillage. Tar or asphalt should never be used near the CLS as they form an explosive mixture when in contact with liquid oxygen.

Sizing

Sizing of the installation is performed by employing a risk assessment (RA) model detailed in Chapter 6 of HTM 02-01 (part A). The RA considers a number of issues amongst which are:

- historic information
- the proximity to the gas supplier
- the potential growth in demand
- average continuous demand (a calculation based on the current clinical demand and areas served)
- vehicular access for delivery tankers.



Figure 1.17 A liquid cylinder installation.

The CLS is normally owned by the gas supply company and the hospital pays a three element charge:

1. The gas price per HCM
2. A service element which includes rental, maintenance and capitalization of the equipment
3. A delivery charge.

Liquid cylinder (LC) installations

Where the annual consumption of a hospital is considered too great for a compressed cylinder manifold but is insufficient for a CLS, then a liquid cylinder (LC) installation can be considered. This type of installation is not dissimilar to a cylinder manifold, having the same configuration of two cylinder banks and a control panel (Fig. 1.17), but has significant advantages in that each LC contains

the equivalent gas capacity of 24 J size compressed gas cylinders. On a typical four-cylinder LC manifold, this is operationally equivalent to 72 size J cylinders (only 75% of the gas capacity is usable as the low liquid level alarm activates when the volume of contents falls to 25%). Another major advantage over compressed cylinders is that the LCs are a semi-permanent installation, and as such are charged from a remote fill point, usually on the outside wall of the compound; this removes the need for manual handling and connecting.

Any pressure build-up in the reserve manifold automatically feeds into the pipeline through the control panel.

The only disadvantage when compared to a CLS is a limitation in overall flow rate. A CLS is capable of 3000 L min^{-1} under normal conditions whereas a LC installation peaks at 500 L min^{-1}. Whilst this is normally sufficient for a mid-range pipeline system, it should be considered when assessing supply options. In general, a LC installation will:

- usually have a 2 × 5 cylinder compressed emergency supply
- have an alarm panel linked to the control panel to provide similar warnings to those of a single CLS installation
- be sited internally or externally with protection from the elements.

OXYGEN CONCENTRATORS (PSA PLANT)

An oxygen concentrator or pressure swing adsorber (PSA) is an alternative source of oxygen utilized when a liquid oxygen supply is either unavailable or impractical, e.g. an off-shore site or where the safety criteria for liquid installations cannot be met.

Operational process

An *oxygen* concentrator operates on the principle of adsorbing under pressure *other* gasses in the atmosphere onto the surface of an adsorbent material, termed a *zeolite*. Oxygen is not adsorbed by the zeolite and passes freely through it into a receiver vessel, ready for later use.

The zeolite is sealed in vessels known as sieve beds which operate in pairs – one adsorbing whilst the other regenerates. The adsorbed gasses, mainly nitrogen, are removed by vacuum pump and are discharged into the atmosphere. The process is capable of producing oxygen concentrations of about 95%, the remainder being made up mainly of argon with a small percentage of nitrogen. This may be of clinical significance as it has been reported that a build-up of argon could occur during closed-circuit anaesthesia.

The major components of a hospital PSA plant are:

- duplex compressors
- receivers
- dryers
- duplex molecular sieves
- vacuum pumps
- filters
- line pressure regulators
- control system
- oxygen performance monitoring system
- a back-up cylinder manifold.

Fig. 1.18 shows a schematic layout for a hospital system.

In addition to the economic costs, a number of other issues must be considered: the process generates a great deal of heat, hence ventilation and cooling for the product and the compressors are major considerations.

Should the plant fail, the emergency cylinder manifold will feed into the pipeline at higher concentrations

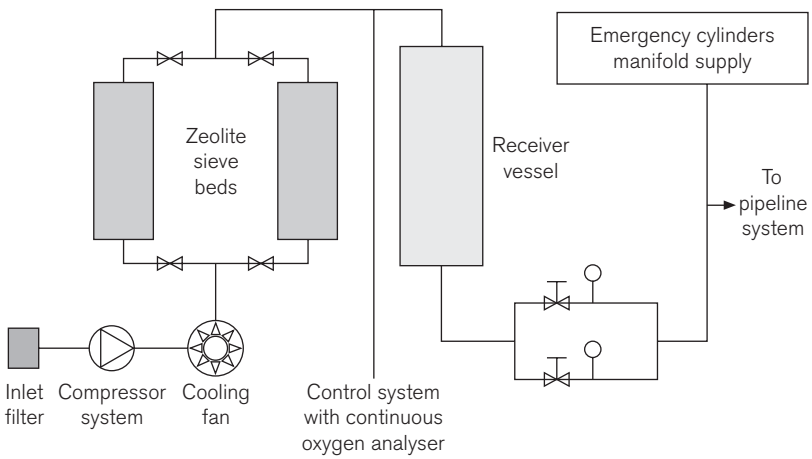


Figure 1.18 A schematic layout for a hospital system.

(99.5%) than the plant's operating norm of 95%. This may have an effect on downstream equipment, particularly in critical care areas.

A more appropriate application for oxygen concentrators is in the home environment where a low-flow, low-pressure system can provide continuous domiciliary oxygen to COPD patients. Typical units (Fig. 1.19) operate on a mains supply and can provide up to 51 min^{-1} at an oxygen concentration of 94%. They are extremely efficient, require little maintenance by the patient apart from periodic cleaning of the inlet filter and can pipe oxygen around the home to small wall-mounted outlets. The concentrator is usually sited in a hallway and needs no special ventilation. The typical noise level when operating normally is around 40 db.



Figure 1.19 A home oxygen concentrator: the Millennium Respironics Oxygen Concentrator.
Image courtesy of Respironics Inc. and its affiliates, Murrysville, PA, USA.

MEDICAL COMPRESSED AIR

Medical compressed air (MA) is classified by the European Pharmacopoeia as a drug, and therefore warrants the same degree of care and cleanliness that any other drug requires during its manufacture, storage or distribution. MA can be provided into pipeline systems by three methods:

1. Compressors
2. Cylinders connected to a manifold
3. By mixing liquid oxygen and liquid nitrogen, also known as synthetic medical air (SMA).

This section will deal with the provision of MA by compressors.

The basic components of MA generation are the same as those found in any industrial compressed air system: a compressor, a receiver for the storage of gas and some form of regulator to monitor and control the pressure in the pipeline (Fig. 1.20). Where MA production differs is in the degree of conditioning applied to the raw compressed air before it is administered to the patient. Compressors for MA plant can be any of three types: reciprocating, Archimedean screw or rotary vane. All have their individual benefits and the ultimate choice of process is based on demand, location and initial cost. Whichever pumps are used they should be identical to each other and capable of meeting total hospital demand with one pump off line.

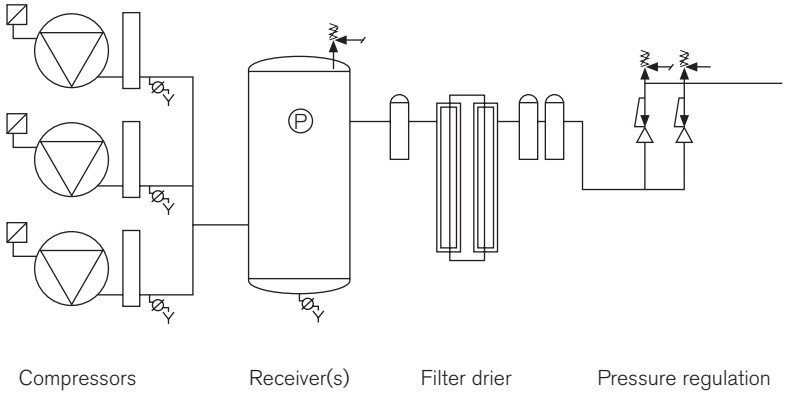


Figure 1.20 Main components of a typical medical compressed air plant.

MA may be contaminated by a number of substances including:

- oil – a lubricating agent used in the compressor that may become an aerosol and be carried into the air stream
- particulate debris – either present in the raw uncompressed air or picked up during compression/storage
- water – a compression process by-product which may collect in the receiver or pipeline.

Regulations mandate the removal of these contaminants before MA is administered to patients; this is achieved by a series of filters and driers installed after compression but before distribution into the pipeline system. See Table 1.10 for the specification required for MA by the European Pharmacopoeia.

The reliable provision of medical air is of critical importance and there are robust mechanisms designed to maintain both the security of supply and the gas pressure. MA is normally distributed at 400 kPa, and is maintained at this pressure by a pressure-reducing valve fitted downstream of the drier/filtration assembly. Further protection is provided by a pressure relief valve that prevents over-pressurization of the system should a fault occur with the reducing valve. Of particular importance is the control system, which integrates all the disparate elements of the supply system, controls the safety back-up devices and ensures that alarms are raised if any component fails to perform in the prescribed manner.

As with all other medical gas supply units, all vital components in the design of the system are provided in at least duplex so that one component (filter/drier/compressor, etc.) can act as a standby if any other component fails or needs to be taken out of service for maintenance.

Medical air is used in patient ventilators, to power humidifiers and drug nebulizers, and in devices such as automatic tourniquets.

Table 1.10 Maximum allowable contaminants for medical air

CONTAMINANT	MAXIMUM ALLOWABLE AMOUNT
Water	60 ppm
Carbon monoxide	5 ppm
Carbon dioxide	500 ppm
Oil	0.1 mg m^{-3}
Sulphur dioxide	1 ppm
Nitrogen monoxide	2 ppm
Nitrogen dioxide	2 ppm

In some areas of the hospital, air is also distributed at a higher pressure for use as a power source for medical tools, such as orthopaedic drills and saws. Known as surgical air (SA), the gas has similar properties to MA, but is delivered at a higher pressure (700–1100 kPa compared with 400 kPa for MA).

Where overall demand for both MA and SA is small, it is acceptable to utilize one compressed air plant to provide both supplies, and to regulate the gas pressures accordingly. Under such circumstances, it is imperative that all regulation and control systems are located in the plant room with both services piped away in parallel from this point. However, some older operating theatres may have a single higher-pressure supply (7 bar) that may be regulated down to 4 bar by plugging in a reducing valve at the wall socket. The Schrader valves used for these sockets must be gas (MA or SA) specific to prevent the wrong supply being used (see below).

Although MA/SA can be provided from manifolds or as synthetic medical air, these approaches are impractical where there are high consumption rates (particularly with SA supplied from cylinder manifolds).

The terminal units will be discussed in detail under pipeline distribution, but it is important to note that those used for both MA and SA are, as with other gasses, of different dimensions, ensuring that the risk of cross connection and delivery of the wrong gas is minimized.

It is sometimes necessary to supply dental or other departments with a sterile supply of air – depending on the distances involved, it may often be more cost-effective to use small local compressors than run the pipework over long distances.

SYNTHETIC AIR SYSTEMS

The 1998 European Pharmaceutical monograph for medical air put greater emphasis on the control of hydrocarbons and moisture in the product and, consequently, an increased need for monitoring the performance of air systems is required. As synthetic medical air (SMA) is produced by mixing liquid oxygen and liquid nitrogen in the gas state, it does not contain either hydrocarbons or moisture and may be considered as an alternative to conventional compressed air supplies, either at the design stage of a new hospital or where purity is an issue. The installation utilizes the existing liquid oxygen supply with an adjoining liquid nitrogen vessel. Both vessels have smaller back-up vessels on the site (Fig. 1.21).

The control system is very similar to the conventional CLS mentioned earlier, with the addition of duplex mixing panels and gas analyzers enabling the continuous monitoring of gas purity. The synthetic air is then stored at either at 4 bar g, or at higher pressures for surgical air and regulated down to 4 bar g for medical use.



Figure 1.21 A synthetic air cryogenic liquid system installation.

The benefits of SMA are that it:

- is maintenance free
- is of high purity
- does not require a power supply
- provides on-site nitrogen for powering surgical power tools.

MEDICAL VACUUM SYSTEMS

Although by definition a vacuum cannot be termed a gas, medical vacuums (MV) are always co-installed with true gas supplies and with the same type of valving and equipment. This, together with the fact that it is covered by the same standards as medical gasses (HTM02, C11 1999 and BS EN ISO 7396), means that piped vacuum services are invariably considered alongside the provision of medical gasses.

The purpose of MV is to enable the removal of fluids during medical or surgical procedures. The principle of fluid removal is the same in all cases: a drainage tube passes from the patient to an interceptor collection jar where any solid and liquid waste is trapped. The vacuum is then passed through a flow regulator and bacterial filter to the terminal unit in the same way as the medical gasses (Fig. 1.22; see also Chapter 20).

The pipeline system carrying the MV back to the plant is usually of the same construction and standard as that of the medical gasses, but of larger size to prevent more significant pressure changes along the pipe. In some instances, pipes over 54 mm can be made of a plastic material to reduce installation costs.

At the vacuum plant (Fig. 1.23), there are additional bacterial traps and collection jars to collect any materials that may have by-passed the patient level filters. Although these components are designed to protect the plant and maintenance staff from contamination, it is advisable to

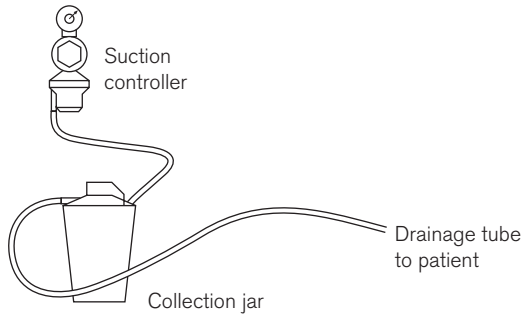


Figure 1.22 Typical components of a ward suction set.

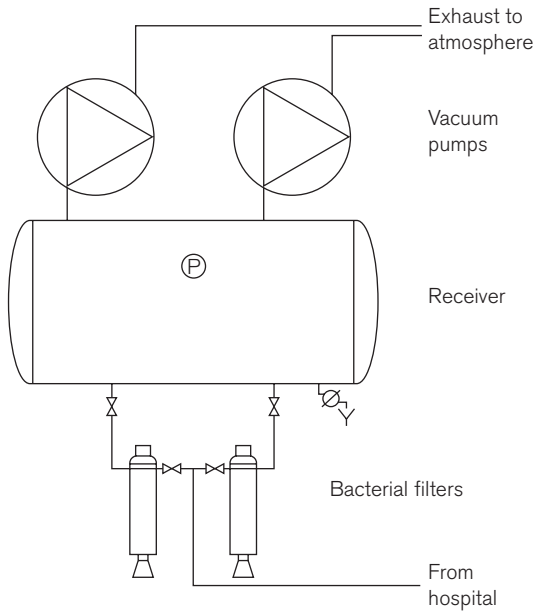


Figure 1.23 Main components of a typical medical vacuum plant.

ensure that strict infection control protocols are adhered to during any maintenance work especially if this involves changing the bacterial filters.

From the filters, the vacuum is drawn into the receiver. The receiver is designed so that the pumps, operating between 550 and 650 mmHg sub-atmospheric, cycle no more than six times per hour. This is good engineering practice and prevents excessive wear and tear on the system.

A MV plant can be thought of as a compressor in reverse; air is taken from the MV pipeline and 'discharged' into the atmosphere. As with medical air, it is essential to have a degree of redundancy in the system. There should be a minimum of three identical pumps with each pump capable of delivering 100% of the design flow rate. Vacuum pump exhausts may be combined, but where this is the case, a non-return valve must be fitted to the exhaust so that it does not 'drive' the standby pump. The exhaust pipeline(s),

however, *must* be vented to atmosphere at high level, normally at roof level and away from all other air intakes or openings into the building (doors, windows, etc.).

The normal pumps used for MV in the UK are of a rotary vane type, although reciprocating pumps are used in some parts of the world. Both of these types of pump have a capacity to generate a sub-atmospheric pressure of up to 650 mmHg at sea level and are perfectly adequate for the purposes of medical vacuum. At higher altitudes though, it is more difficult to achieve the negative pressures required and the settings on the plant control systems need to be adjusted to compensate for this lower operating range.

Again, as with the compressed air plant, the most important element is the plant management and control system. This operates the cut in and cut out of the pumps, cycles the pumps on duty (so that each pump experiences the same amount of use) and passes any faults back to the alarm and indication system.

In certain parts of the world (e.g. the US) medical vacuum systems are also used to deliver the negative pressure requirements for anaesthetic gas scavenging system (AGSS). In the UK we tend to utilize a totally separate vacuum source for this purpose; a less powerful vacuum but with higher flow rates (120 L min^{-1} per terminal unit, as opposed to 40 L min^{-1} for MV). As this vacuum source is of a lower technical specification, greater savings can be made both in capital terms and in running costs.

Performance levels and specifications for a medical vacuum service

For the UK, the specifications for a piped vacuum service are laid down in the HTM 02-01:2006. In brief the guidance states that:

- the design and operating pressure should not be less than 450 mmHg at the plant
- a pressure drop of 50 mmHg within the distribution pipework is permissible
- a minimum pressure of 400 mmHg is required at the back of the terminal unit
- a pressure drop of 100 mmHg is allowed across the terminal unit to the probe, which has to maintain a minimum pressure of at least 300 mmHg whilst delivering a flow rate of 40 1 min^{-1}.

ANAESTHETIC GAS SCAVENGING SYSTEMS

These are considered specifically in Chapter 18. They do, however, form part of what is termed Medical Gas Piped Services, and common aspects of the piping and distribution are considered further in this chapter.

ALARM AND INDICATION SYSTEMS FOR PIPED GASSES

There are two different types of alarm system used within a hospital medical gas system: main plant alarms and local (or ward) alarms. The former is used to provide an indication of the condition of the plant at the source of generation or storage, the latter to provide an indication of the condition of the gas at the point of use.

The main plant alarm (Table 1.11) consists of a series of panels placed in strategic locations throughout the hospital. These will usually give the indication that everything is normal; their main function though is to give advance

Table 1.11 Main plant alarm

SERVICE LEGEND	LIQUID OXYGEN	MANIFOLD SUPPLY	AIR PLANT	VACUUM PLANT
Normal	Normal	Normal	Normal	Normal
Condition 1	Refill liquid oxygen	Change cylinder	Plant fault	Plant fault
Condition 2	Refill oxygen immediately	Change cylinder immediately	Plant emergency	Plant emergency
Condition 3	Reserve low	Reserve low	Reserve low	
Condition 4	Pressure fault	Pressure fault	Pressure fault	Pressure fault

warning of a potential system failure. For example, if the duty bank on a manifold runs out, the standby bank will automatically come on-stream. As soon as this happens, the *first condition alarm* will be triggered indicating that cylinders need changing on that manifold. The service is not in danger, as the manifold is designed to act in this way. If no one attends to the manifold and the standby bank also runs out, the *second condition alarm* will be activated: at this point the system is about to run out of gas. If the pressure does fall below the minimum required then the final condition – *pressure fault* – will commence. At this stage, patients will need to be provided with alternative supplies.

The third condition on the system is used to monitor the failsafe emergency supply source; although this should not be used as a main supply, it may provide the hospital with enough time to rectify matters.

As well as the indications on the main alarm panels, additional indications will appear on each plant control panel or manifold. These provide a more detailed visual indication of the nature of the fault or emergency.

A local or area alarm panel fulfils a very different function. Here the alarm condition is used to indicate that something has *already* gone wrong. Each gas supplied to a ward or department is monitored for faults by a pressure switch mounted in the pipeline, downstream of the final area valved service unit (AVSU). Typically this is set at ±20% of the line pressure specified for a particular gas such that, if a high- or low-pressure condition occurs within the area, the alarm will indicate the fact (Table 1.12).

On both types of alarm panel the indication is both audible and visual. A two-tone sounder and a flashing legend indicates what the fault is and on what service. The audible alarm can be muted but will reinstate itself after 15 minutes if the fault has not been corrected. On clearance of the fault, the alarm panel will automatically reset itself to 'Normal'.

Table 1.12 Typical local alarm panel legends

SERVICE LEGEND	OXYGEN	MEDICAL AIR 400 KPA	VACUUM
Normal	Normal	Normal	Normal
Condition 1	High pressure	High pressure	High pressure
Condition 2	Low pressure	Low pressure	Low pressure

DISTRIBUTION SYSTEMS

Medical gasses (other than surgical air) are distributed throughout the hospital at a nominal 400 kPa through pipelines designed to minimize the pressure drop from source to point of use. This is achieved by means of calculations based on the initial pressure, the specified flow rate and the dimensions of the pipework. In simple terms, the higher the required flow rate, the larger the diameter of the pipe needed to carry it. So, for example, a high flow pipeline in a plant room can be 54 or even 76 mm diameter, whereas by the time the pipes enter the ward they normally do not exceed 22 mm and those supplying the patient's bedside, are usually 12 mm.

The 'gasses' normally distributed by pipeline in hospitals within the UK are:

- Oxygen O_2 (400 kPa)
- Nitrous oxide N_2O (400 kPa)
- Entonox 50% O_2/50% N_2O (400 kPa)
- Medical air MA (400 kPa)
- Surgical air SA (700 kPa)
- Vacuum vac
- Anaesthetic gas scavenging AGSS.

They all carry an individual colour code, as shown in Figure 1.24.

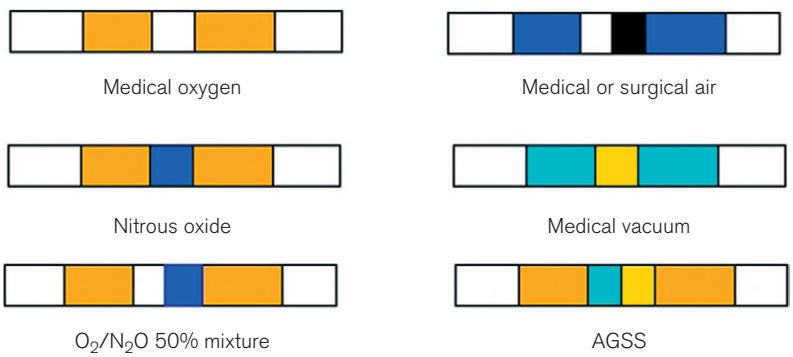


Figure 1.24 Colour codes for medical gas pipework.

Other gasses, such as helium and hydrogen, are usually only supplied as piped services to pathology laboratories and so are not considered further.

Detailed information concerning the regulations and standards required for fixed distribution pipework can be obtained from the appropriate Government or Health Ministries. In the UK, the ordinances covering the service, maintenance, repair or alterations of fixed distribution pipework systems are detailed in HTM 02-01.

In this chapter, only a brief description will be given of the fixed pipework, as it resides 'behind the wall' and is more appropriately the concern of the hospital engineer. The anaesthetist or designated medical officer should, however, be aware of the nature of the installation and should always be informed and consulted before any alterations to it are made.

The pipes used are *half hard* and manufactured from phosphorous de-oxidized non-arsenical copper to BS EN 1412:1996 grade CW024A to prevent degradation of gasses. 'Half hard' refers to the heat treatment of copper pipes, which allows them to have a higher pressure rating. Pipes are degreased, purged, filled with nitrogen and capped to maintain cleanliness prior to delivery and installation. Pipefittings used for jointing these pipes are made from the same materials.

Valves need to be installed at various points along the network: on exiting the plant room, entering buildings, at the branch of each riser and on entry to each department or ward.

Valves within plant rooms should be left unlocked but all other valves should be locked and only unlocked under a permit to work order and the supervision of the 'authorized person'.

Valves at ward entrances or departmental isolation valves are normally termed AVSUs (Area Valved Service Units) or ZSU (Zone Service Units) and are specifically designed to provide not only gas isolation but also other functions as described below. Essentially, an AVSU comprises a lockable box that has a glass-fronted panel, which can be broken by the ward staff to allow isolation of gas flow to areas in case of fire, pipeline fracture or other emergency (Fig. 1.25).

AVSUs also permit additional connections into the gas stream by the use of a non-interchangeable screw thread (NIST) system. The NIST union contains a self-sealing valve so that when the blanking nut is removed and the appropriate NIST connection is made, the self-sealing valve is automatically reopened. These branches may be used to purge pipelines or to introduce a local supply during alterations or breakdowns. The AVSU junction also allows the insertion of a 'spade', which is used by service engineers to ensure absolute closure of the pipeline, irrespective of the action of the valves.

AVSUs are installed in such positions as to protect each department or ward. They should be installed into the natural route that would be passed on an emergency exit from the department unit.



Figure 1.25 A typical area valved service unit (see text), note blanked off non-interchangeable screw thread connectors on either side of valve.

The pipework itself should be identified by labels placed upon it at regular intervals (Fig. 1.24) in accordance with the identification code described in Section 13 of HTM 02-01 and further marked as to the direction of flow.

Pipework is normally always concealed in modern-day installations, though in the past it was mounted on the surface. The older arrangement was not only unattractive, but also less satisfactory from the standpoint of general hygiene and cleanliness.

Terminal outlets

The distribution pipework terminates in wall- or pendant-mounted self-sealing socket outlets (Fig. 1.26).

The specification for the design of terminal units and their probes was originally set in 1978 (BS 5682) and upgraded in 1998 (BS EN 739:1998 'Low pressure hose assemblies for use with medical gasses') – this is now in the process of being superseded by a new standard: BS EN ISO 9170–1:2008. Current legislation specifies that the terminal unit should consist of two sections:

1. A termination assembly permanently attached to the appropriate pipeline
2. A socket assembly containing the quick connect probe socket, commonly known as a *Schrader socket* in the UK. This can be removed by a service engineer, but must be designed so that it cannot be accidentally connected to a different gas service.

A termination assembly for a pressurized gas (but not a vacuum) must also have a check valve so that work can be carried out on any terminal unit without shutting down all the terminal units for that gas in that area. The valve should operate automatically as soon as the socket assembly is removed.



Figure 1.26 Terminal outlets. Note the different diameter recesses (collar indexing system) that match the collar on the relevant probe (see Fig. 1.27).



Figure 1.27 Flexible hose probes. Note the difference in size of the indexing collar.

It is essential that the assembly design ensures gas specificity for each component, such that no terminal unit can be assembled and include parts from different gas termination outlets. The identity of the gas for each terminal unit should be permanently displayed on all individual components. The socket assembly, when assembled will only accept a probe with the same gas identity, by utilizing a collar indexing system that is unique to that gas service.

Flexible pipeline

This connects the terminal outlet to the medical equipment; it has three components:

1. A quick connect (Schrader) probe that fits the terminal outlet
2. A flexible hosepipe
3. A NIST connection that fits the equipment, e.g. the anaesthetic machine.

Quick connect probe

The male part of the probe for the terminal outlet is the same size for all gasses, but in order to prevent connection to the wrong gas service it has a protruding (gas specific) indexing collar (Fig. 1.27). This collar has a unique diameter that only fits the matching recess of the socket assembly for that gas.

The British Standard also stipulates the following:

- It must not be possible to twist the probe while it is connected to the unit (unless it is connected vertically to a pendant). To this end, the collar is provided with a notch that fits over a rigid pin in the socket assembly.
- It must be possible to insert or remove the probe simply and quickly using only one hand. In order to achieve this, the socket assembly has a spring-loaded outer ring, which when depressed releases the locking mechanism holding the probe, and causes the probe to be ejected.
- The unit must seal off the flow of gas when the probe is withdrawn.

The flexible hosepipe

This is the section of the system in which damage and wear are most likely to occur. Originally, the hoses for each gas were constructed from the same black reinforced rubber or neoprene tubing, and only identifiable by a short length of coloured sheath at each end. This resulted in several fatal incidents when the probe for one gas was fitted to the upstream end of the hose, but a socket or union for a different gas was attached to the downstream end. The incidence of such accidents has been minimized by the current practice of most manufacturers, which is:

- to produce complete hose assemblies (with the appropriate connections pre-attached) for the different gas services, under strict quality-control systems and with necessary checks to ensure correct application. Furthermore, it is now recommended practice that a damaged hose should not be repaired on-site but returned in its entirety to the manufacturer in exchange for a factory-made service replacement
- to use characteristically coloured tubing for each gas (Fig. 1.28).

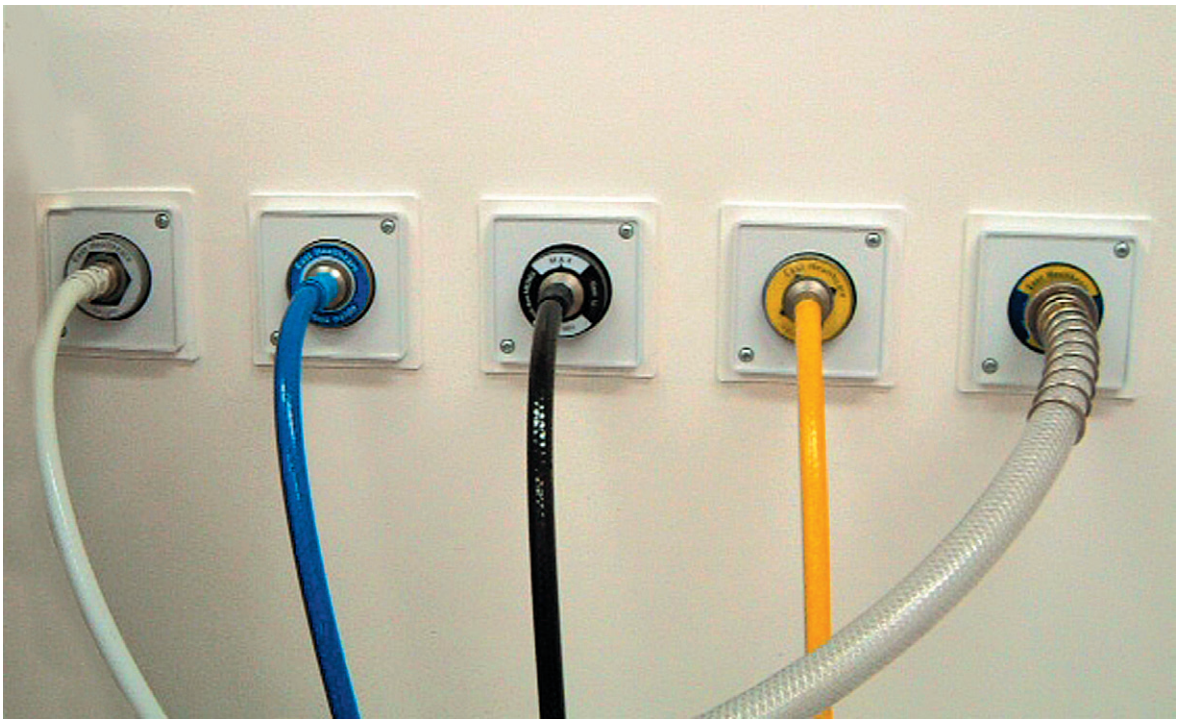


Figure 1.28 Colour-coded hoses.

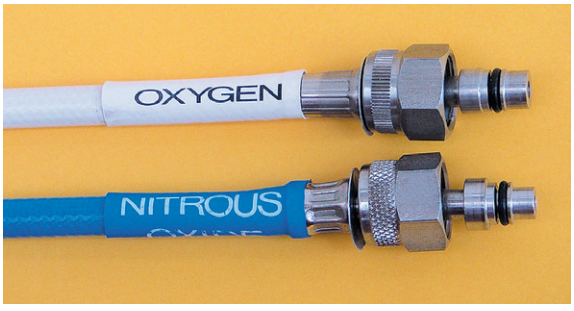


Figure 1.29 Non-interchangeable screw thread connectors with the locking collars manipulated to reveal the different gas-specific profiles of the probes.

The non-interchangeable screw thread (NIST) connector

To ensure that the gas supply is attached correctly to the relevant piece of medical equipment, each hose is fitted with an unique downstream connector. This takes the form of a probe and nut (Fig. 1.29). The probe has a unique profile for each gas supply and fits only the union on the receiving equipment. This profile consists of two cylindrical shapes which together form a unique combination. For example, in Fig. 1.29 the outer part of the nitrous oxide probe is smaller than that for oxygen, but the inner part is larger. Thus, although the outer tip of the nitrous probe fits into the oxygen receiver, the larger inner part will prevent full engagement. Conversely, the outer part of the oxygen probe is too large to engage with the nitrous oxide receiver.

The nut has the same diameter and thread (in the UK) for all the gas services, but can only be attached to the anaesthetic machine when the probe is correctly engaged. The term 'non-interchangeable screw thread connection' is ambiguous as it can give the impression that the screw threads are different and cause the unique fit; however, the gas specific fit is actually dependent on the collar configuration.

The connections between the hose and fittings must be secure and tamper-proof. Both the British Standard (Schrader) terminal units and NIST probes have serrated spigots, which are pushed into the ends of the hosepipe. To prevent their working loose, a stainless steel sleeve (ferrule) is placed on the outside of the hosepipe and spigot and compressed (crimped) by a 30-ton press. The ferrule is sufficiently robust to defy all but the most determined attempts at removal and compresses the hosepipe onto the spigot with such force that any attempt to pull the two apart will result in the hose stretching and breaking before the connectors are pulled off.

TESTS AND CHECKS FOR MEDICAL GAS PIPED SERVICES

- Anaesthetists are only responsible for checking that part of the medical gas pipeline services (MGPS) system between the terminal unit and the patient. They should be able to take the quality and unfailing supply of gasses for granted.
- Quality control is usually considered to be the province of the hospital pharmacist, who should order, or make, tests to confirm the identity of the gas, its purity and composition, and freedom from contaminants, including solid particulate matter. Compressed air should also be examined for water vapour and oil mist on a quarterly basis to ensure compliance with the European Pharmacopoeia. The pharmacist is also usually responsible for maintaining adequate supplies of cylinders.
- The engineering department under the direction of the authorized person (AP) is responsible for the day-to-day management and operation of the MGPS and for organizing planned preventive maintenance, emergency repairs, design compliance, continuity of supply and indication/warning systems (alarms).
- Designated theatre staff are usually responsible for changing cylinders on anaesthetic machines and maintaining a store of portable oxygen cylinders with flowmeters and suction equipment for use in emergencies or during shut down for maintenance and alterations.
- The anaesthetist is responsible for the correct insertion of the pipeline probes and any necessary adjustments.
- HTM 02-01 (part B) describes a 'permit to work' system. Essentially, this is a code of practice for repairs and preventive maintenance on the MGPS system in which the engineer discusses with the appropriate people the nature and timings of the work to be done and ensures that independent services for medical gasses such as oxygen and vacuum are made available as required. The 'permit to work' document has five parts and may at first seem to be yet another proliferation of the already burdensome paperwork in hospitals. It does, however, increase safety and improve the relationships between departments.
- Finally, and most importantly, the whole operation and management of the medical gas system is described in HTM 02-01 part B – Operational Management. This requires that there be a written policy on the use and management of medical gasses with defined responsibilities and roles. It is essential that this document is produced for each hospital, specific to the risks and procedures of that particular site. The medical gas supplier should be in a position to assist in its production.

In order to comply with HTM02-01 and current Health and Safety at work legislation, it is essential that all staff who are responsible for handling or administering medical gasses are appropriately trained in their safe use, and are aware of the potential risks and dangers. A number of medical companies such as BOC provide courses covering these issues.

FURTHER READING

BOC Gas. *Safe with medical gasses*. Guildford, Surrey: BOC Ltd; 2002.

British Standards Institute. *Part 1 Medical Gas Pipeline Systems – Terminal Units for Compressed Medical Gasses and Vacuum*, BS EN 737. Milton Keynes, UK: British Standards Institution; 1998.

British Standards Institute. *Anaesthetic and Respiratory Equipment Compatibility with Oxygen*, BS EN ISO 15001. Milton Keynes, UK: British Standards Institution; 2004.

British Standards Institute. *Part 2 Medical Gas Pipeline Systems – Anaesthetic Gas Scavenging Systems – Basic Requirement Medical Gas Pipeline Systems*, BS EN 737. Milton Keynes, UK: British Standards Institution; 1998.

British Standards Institute. *Part 3 Medical Gas Pipeline Systems – Pipelines for Compressed Medical Gasses and Vacuum*, BS EN 737. Milton Keynes, UK: British Standards Institution; 2000.

British Standards Institute. *Part 1 Pressure Regulators for use with Medical Gasses – Pressure Regulators and Pressure Regulators with Flowmetering Devices*, BS EN 738. Milton Keynes, UK: British Standards Institution; 1997.

British Standards Institute. *Part 2 Pressure Regulators for Use with Medical Gasses – Manifold and Line Pressure Regulators*, BS EN 738. Milton Keynes, UK: British Standards Institution; 1999.

British Standards Institute. *Part 3 Pressure Regulators for Use with Medical Gasses – Pressure Regulators with Integral Cylinder valves*, BS EN 738. Milton Keynes, UK: British Standards Institution; 1999.

British Standards Institute. *Specification for Probes (quick connectors) for use with Medical Gas Pipeline Systems*, BS 5682. Milton Keynes, UK: British Standards Institution; 1998.

British Standards Institute. *Transportable Gas Cylinders.Gas Cylinder Identification (Excluding LPG) Colour Coding*, BS EN 1089–1083. Milton Keynes, UK: British Standards Institution; 2004.

HTM 02-01. *Medical Gas Pipeline Systems – Design, Installation, Validation and Verification*. London: The Stationery Office; 2006.

Grant WJ. *Medical gasses – their properties and uses*. Aylesbury: HM & M; 2005.

International Organization for Standardization. *Small Medical Gas Cylinders, Pin Index Yoke-type Valve Connections*, ISO 407. Geneva: International Organization for Standardization; 2004.

International Organization for Standardization. *Cylinder Valve Outlets for Gases and Gas Mixtures – Selection and Dimensioning*, ISO 5145. Geneva: International Organization for Standardization; 2004.

International Organization for Standardization. *Gas Cylinders for Medical Use – Marking for Identification of Content*, ISO 32–31. Geneva: International Organization for Standardization; 1977.

International Organization for Standardization. *Terminal Units for Medical Gas Pipeline Systems – Part 1: Terminal Units for Use with Compressed Medical Gases and Vacuum*, ISO 9170–9171. Geneva: International Organization for Standardization; 2008.

Chapter 2

Measurement of pressure and gas flow

Paul Beatty

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Any system for delivering anaesthetic gasses requires the consistent and accurate measurement of gas flow and pressure. In order to be able to understand the detailed design of modern anaesthetic machines, this chapter will consider how such measurements are obtained and the basic physical principles involved.

FORCE, PRESSURE AND FLOW

Force is that which causes an object to move or, more accurately, to accelerate. Once it is moving, Newton's Laws tell us that it will continue to move in a straight line at a constant rate unless some other force is applied to it. The SI unit of force is the *Newton (N)*, which is defined as the force that causes a mass of 1 kg to accelerate by 1 metre per second per second ($m\ s^{-2}$). Other units of force occasionally used are the kilogram force (kgf) and the pound force (lbf). These are forces equivalent to the force exerted by the earth's gravity on masses of a kilogram and a pound, respectively.

Pressure is force per unit area over which the force acts, i.e.

$$\text{Pressure} = \text{Force}/\text{Area}$$

In any gas or liquid, pressure acts in all directions equally, whereas force acts in a given direction. For example, in a full hypodermic syringe, the liquid can be pressurized by applying a force to the plunger of the syringe. This is applied in the direction of travel of the plunger. However, if there were a leak in the barrel of the syringe, the liquid would squirt out sideways from the leak, as well as from the syringe outlet (Fig. 2.1), due to the pressure created by the force acting in all directions within the barrel of the syringe. The amount of pressure generated depends on the area of cross section of the barrel since it is over this area that the force acts. Thus, the pressure generated in a syringe with a small bore is higher than that generated in a syringe with a large bore for the same force applied to the plunger.

The SI unit of pressure is the *Pascal (Pa)*, which is the pressure that results from the application of 1 N over an area of 1 m^2. Because a pressure of 1 Pa is rather small, gas pressures in anaesthesia tend to be measured in kPa. Other units of pressure that may be used in anaesthesia are pounds per square inch (psi) and bar (the pressure exerted by the atmosphere at sea-level, which is the result of the weight of atmospheric gasses bearing down on the surface of the earth).

Flow can result from differences in pressure. If a liquid or gas encounters a region in which one point has a higher pressure than another, then it will move away from the point of higher pressure towards that of lower pressure. That motion is *flow*. The commonest type of unit of flow encountered in anaesthesia considers the volume of flow of a liquid or gas in a given time. The SI unit is metres cubed per second ($m^3\ s^{-1}$). As this is very unwieldy, litres per second ($l\ s^{-1}$) or litres per minute ($l\ min^{-1}$) are normally used. Strictly, these units describe *Volume Flow* where it is assumed that that pressure and temperature are constant. It is useful to have a measurement of flow that is independent of pressure and temperature.

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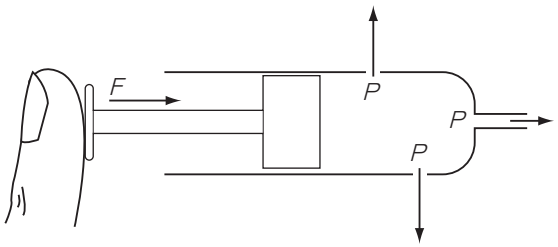


Figure 2.1 When the user exerts a force *F* on the plunger of an intravenous syringe, it is applied in one direction only, but the resulting fluid pressure *P* is exerted in all directions.

All the individual gas laws, i.e. Boyle's Law, Charles's Law and Gay Lussac's Law, can be summarized in the equation below using the *Ideal Gas Law* which relates the pressure and volume of a given mass of gas to its temperature:

$$PV = nRT$$

where *P* is the pressure of the gas, *V* its volume, *T* its temperature, *R* a constant and *n* the number of moles of gas present. Measuring the change in the number of moles (*n*) of a gas in a stream gives a measurement of the number of atoms moving in the flow and is called *Molar or Mass Flow*.

ATMOSPHERIC PRESSURE AND PARTIAL PRESSURE

The definition of *bar* given above shows that the air exerts a pressure called *atmospheric pressure*. Atmospheric pressure is measured using a barometer, the simplest form of which is the Fortin barometer (Fig. 2.2). This consists of a long transparent tube, sealed at one end, which is filled with mercury and inverted with its open end in a bath of mercury exposed to atmospheric pressure. The atmospheric pressure acting on the surface of the mercury in the bath will support a column of mercury of about 760 mm above the surface of the mercury in the bath, leaving a virtual vacuum between the surface of the mercury in the tube and the sealed tube end. The height of column of mercury is measured using a Vernier scale at the top of the tube, which can be adjusted using a knob next to the barometer tube.

This measurement of atmospheric pressure leads to a further unit for pressure, millimetres of mercury (mmHg or Torr). If the tube of the barometer is 1 cm^2 in diameter, a column of mercury 760 mm high will weigh 1.033 kg, which means that atmospheric pressure is 760 mmHg = 1 bar = 1000 millibars = 15 psi = 1.033 kg cm^{-2} = 1.01×10^5 Nm^{-2} = 101.3 kPa.

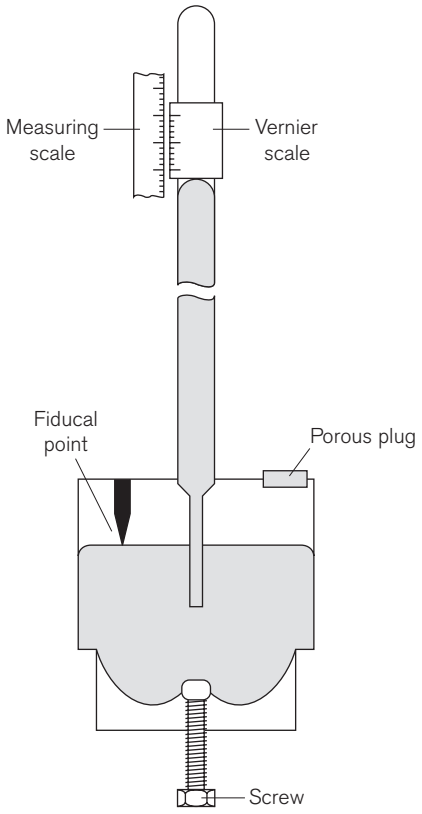


Figure 2.2 Fortin barometer. The level of the surface of mercury in the lower chamber is equilibrated to ambient pressure via the porous plug. The screw is adjusted until the surface of the mercury is touching the fiducal point. The level of the mercury in the column then represents the ambient pressure and is measured using the Vernier scale.

Partial pressure

Returning to the *Ideal Gas Law* (see above), it is clear that it makes no difference whether different gasses contribute the total number of moles present in any mixture of gasses. The Ideal Gas Law for a mixture of two gasses can be rewritten as:

$$PV = (n_1 + n_2)RT = n_1RT + n_2RT$$

This is *Dalton's Law of Partial Pressures* and shows that each gas exerts its own *partial* pressure independent of its companions. For example, if air is 20% oxygen and 80% nitrogen, then for an atmospheric pressure of 760 mmHg, the partial pressure of oxygen will be 152 mmHg and the partial pressure of nitrogen will be 608 mmHg. However, as Figure 2.3 shows, barometric pressure falls with altitude. Thus, when treating a patient at high altitude, a larger percentage of oxygen is required to supply the same

partial pressure of oxygen. It is this partial pressure, not percentage of gas, which is clinically most important. The relationship between altitude and the partial pressure of the components of atmospheric air is shown in Figure 2.4.

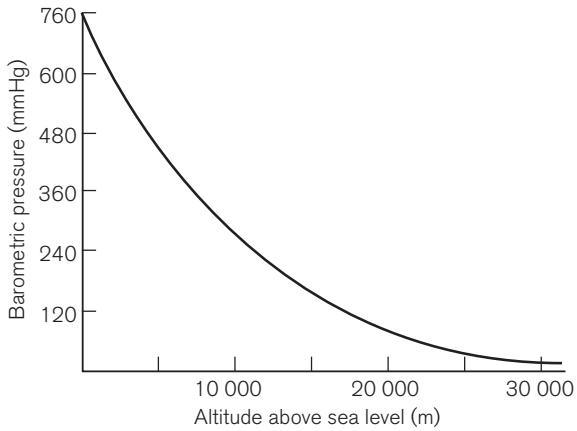


Figure 2.3 The relationship between barometric pressure and altitude above sea level.

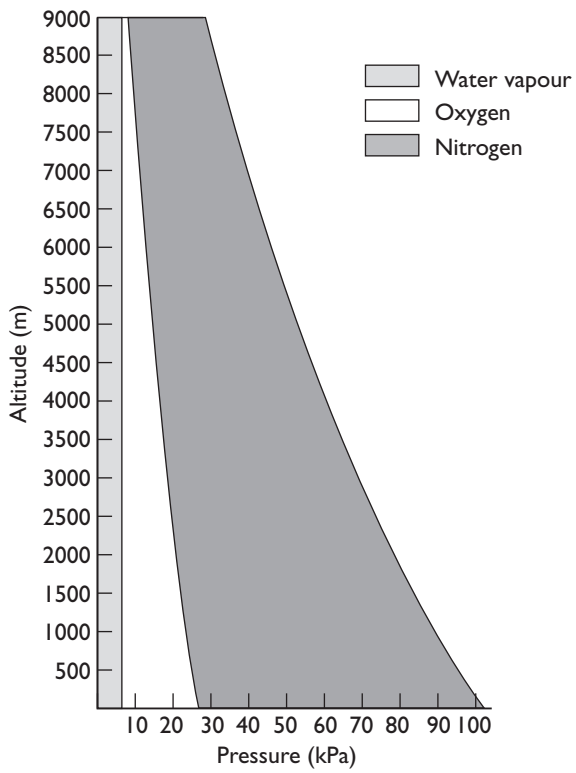


Figure 2.4 Variation with altitude of components of inspired air. Saturation with water vapour by upper airways is assumed.

ABSOLUTE, DIFFERENTIAL AND GAUGE PRESSURES

Since the barometer measures pressure with reference to effectively a vacuum, the measurement made is an *absolute pressure measurement*. There are other types of absolute measurements of pressure made with anaesthetic equipment. Mostly these are related to measurements of high pressures such as those encountered in gas cylinders.

Differential pressure measurement is used where the difference in pressure between two points is required. The simplest differential pressure measurement system is the manometer, as shown in Figure 2.5. This takes the form of a U-shaped tube, partially filled with liquid such as water. Gas at pressure *P* is applied to end A, with end B open to the atmosphere (so having atmospheric pressure applied to it). The pressure at A causes water to be pushed to the other limb of the U so that the level in the right-hand limb moves to point *b* above where it was before the pressure was applied, and the water in the left-hand limb is depressed below its original level. If both move by a distance of 5 cm then the difference in water level created is 10 cm, indicating a pressure difference between A and B of 10 cmH_2O. Since the end B is open to atmosphere, the pressure at A can be thought of as being above atmospheric and is referred to as *gauge pressure* and is sometimes denoted by placing a '**g**' after the pressure unit, e.g 10 cmH_2O g. If the ends A and B are at different points within a closed system, such as between the pleural cavity

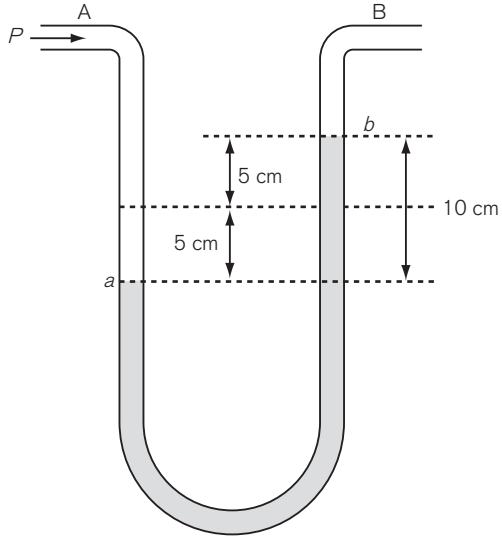


Figure 2.5 A simple manometer. A pressure *P* is applied at A. This causes a depression of the fluid level at *a* and a corresponding rise of the fluid level at *b*. In this case the tube is filled with water and the pressure is 10 cm of water.

and the alveoli, then the pressure measured is a true differential pressure.

METHODS OF MEASURING PRESSURE

Mechanical methods

Bourdon gauge

The Bourdon gauge (Fig. 2.6) is robust, inexpensive and can withstand high pressures. It consists of a curved flattened tube, elliptical in section. When pressure is applied, the tube expands and in doing so attempts to straighten out. Levers, gears or a rack and pinion mechanism translate this movement to a dial pointer. The inlet has a constriction within it to protect the gauge from sudden increases in applied pressure. This gauge is normally used in anaesthesia to indicate cylinder and pipeline pressures. In this application, the pressures measured are much greater than atmospheric pressure. As the response to the applied pressure is determined by the mechanical properties of the Bourdon gauge tube, it effectively measures absolute pressure.

Aneroid gauge

The mechanical principles of an aneroid gauge are shown in Fig. 2.7. It measures absolute pressure and, as with the Bourdon gauge, the movement generated by application of pressure to a chamber is translated into movement of a dial by a mechanical linkage. The amount of movement generated is controlled by the compliance of the aneroid chamber so that the more rigid the chamber the higher the pressure that the gauge will indicate. The sensitivity of the gauge can be controlled by the gearing ratio of the rack and pinion. Where encountered in anaesthetic devices, aneroid gauges are relatively delicate and sensitive but able to indicate low pressures. They may be used to measure airway pressure or blood pressure. If the chamber is sealed and evacuated, the gauge becomes the familiar *aneroid barometer.* Electronic sensors of pressure have largely superseded the aneroid gauge.

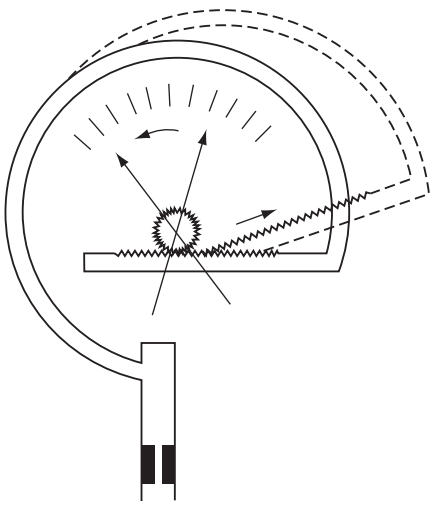


Figure 2.6 The principle of the Bourdon tube pressure gauge. Note the constriction at the inlet.

Electronic methods

Solid-state electronic pressure transducers

Electronic pressure gauges (transducers) are now the commonest method for the measurement of pressure and (in modified forms) force, in anaesthetic machines and devices such as blood pressure machines or infusion pumps. They can be used for both absolute measurements and differential measurement, depending on how they are housed and mounted. A typical solid-state electronic pressure transducer is shown in Fig. 2.8.

When used as an absolute pressure monitor, the transducer is mounted within a chamber and is isolated from the medium in which the pressure is to be measured by a flexible membrane such as a colloid gel to prevent contamination or loss of performance caused by the medium. The sensitivity and pressure measurement range of the transducer is determined by the inherent sensitivity of

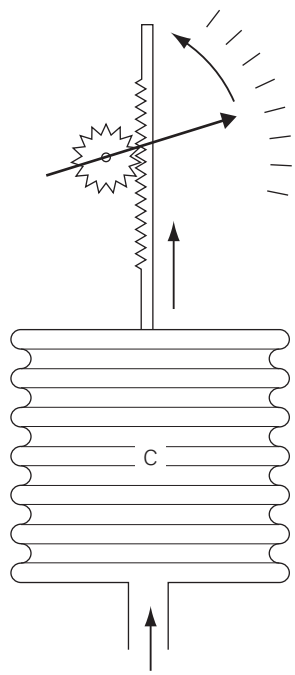


Figure 2.7 Low-pressure aneroid gauge with low-pressure chamber (C).

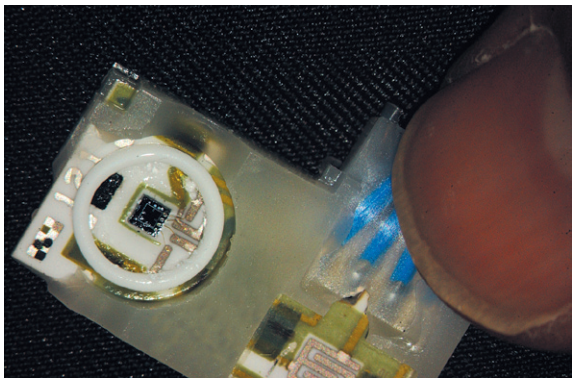


Figure 2.8 A solid-state electronic pressure transducer as used in a typical single-patient use invasive arterial pressure monitoring set. The coupling gel or 'membrane', which is interposed between the lumen of the monitoring line and the transduction chamber, has been removed; remnants of this remain visible. A ceramic plate, onto which is printed the electrical circuit, houses the piezoresistive chip which forms the base of the chamber. An 'O' ring seals the chamber against the housing of the transducer.

the *piezoresistive* transducer (see below). If used as a differential pressure transducer, the sensor is mounted so that each side of the transducer is connected to one of the two possible pressure inlets.

Piezoresistivity is the property of a material whereby its electrical resistance changes when subjected to mechanical stress. Although this is a common property, semiconductor materials show particularly large piezoresistive responses and are therefore used to make very sensitive pressure transducers (strain gauges).

In a solid-state pressure transducer, a single piezoresistive strain gauge is formed on a thin single piece of silicon. The resistance of this strain gauge becomes higher or lower as the silicon slice is flexed by the applied pressure. The resistor is used as one of the four resistors of a Wheatstone bridge circuit (Fig. 2.9), where four resistances are placed in a diamond formation. The voltage difference across the bridge is zero, that is, the bridge is balanced when:

$$\frac{R_1}{R_2} = \frac{R_3}{R_4}$$

When out of balance, the voltage generated across the bridge can be arranged to be proportional for small changes in the resistances, provided the bridge is near to balance. Strain gauges in general, and piezoresistive gauges in particular, are very sensitive to changes in temperature. Provided all the resistances within the Wheatstone bridge are exposed to the same temperature changes, all the resistances change proportionally and thus the balance condition is unaffected. In solid-state pressure transducers, it is relatively simple to ensure that all the resistors in the bridge are exposed to the same temperature changes, by placing them in the same silicon slice that contains the strain gauge but in a position where they are not subject to mechanical stress.

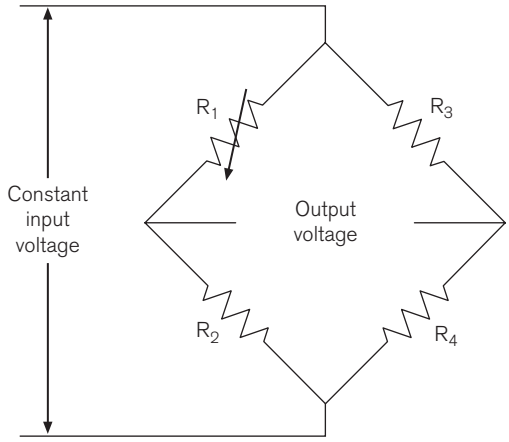


Figure 2.9 A Wheatstone bridge circuit. In the Wheatstone bridge the output voltage is balanced and equal to zero when $R_1/R_2 = R_3/R_4$ for a constant input voltage or current.

MEASUREMENT OF GAS FLOW

The measurement of gas flow is vital to anaesthesia and can be approached using a number of different physical principles. The practical flowmeters include those that respond to gas velocity, volumetric flow, gas momentum and mass flow. All are affected by whether the flow to be measured is laminar, turbulent or reciprocating. In some types of flowmeter, the design of the flowmeter itself ensures that the required flow regime is present, in others the operating flow conditions are set by the application. In addition to being sensitive to the flow regime, they are also affected by changes in gas composition. The latter makes the measurement of respiratory flows, where expiratory and inspiratory gas composition and humidity vary, inherently more difficult than the measurement of flow in dry, single-gas composition situations within anaesthetic machines. The correct selection of the type of flowmeter for the chosen application is thus essential. Table 2.1 shows the general characteristics upon which selection of a type of flowmeter depends.

Differential pressure flowmeters

Constant area differential pressure flowmeters

Three types of differential pressure flowmeter are in common use (Fig. 2.10). In all three, some sort of resistance to flow is placed in the flow stream and the pressure drop across the resistance is calibrated as a measurement of flow.

Table 2.1 Characteristics of flow and volume measurement in anaesthesia

Gas flow *into* breathing system
Continuous flow
Dry gas
Resistance to flow irrelevant
Slow response to change in flow rate
Single or mixed gasses
Gas flow and volume *within* breathing system*
Intermittent flow
Wide range of flow rates
Rapid response
Very low resistance to flow
Mixed gas composition
High humidity
Integration of flow, breath by breath

*NB *Exhaled* volumes should *always* be measured rather than inspiratory volumes, as then the anaesthetist always has an indication that at least this volume was previously inhaled despite any unanticipated leaks in the system.

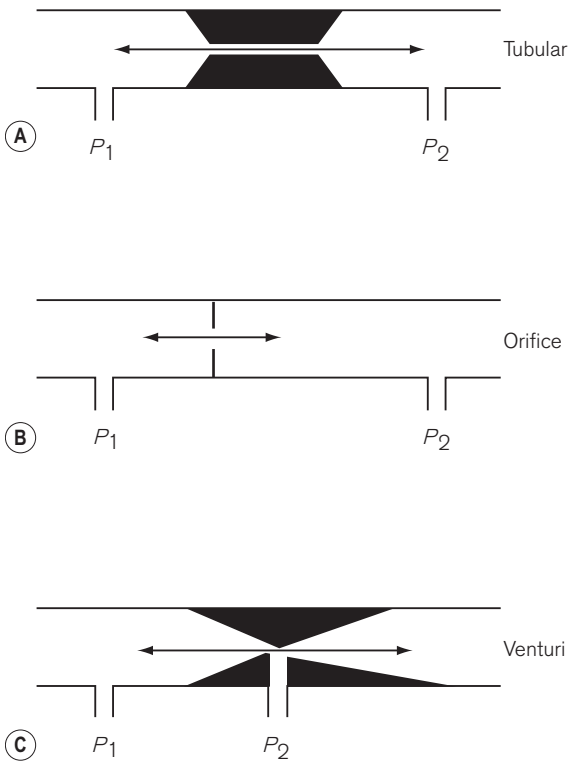


Figure 2.10 Differential pressure flowmeters. P_1, P_2, differential pressures.

In the case of the *tubular flowmeter* (Fig. 2.10A), the pressure drop across a tube (or set of tubes) in the stream is measured. Provided the tube is long enough, this causes even turbulent flow to become laminar. As a result, the pressure drop P across the tube is proportional to the volumetric flow rate F, and the constant of proportionality is the resistance R:

$$P = RF$$

Resistance is inversely proportional to the area of the tube and proportional to its length. Since tube length is usually limited by practical considerations and the need to make the flow in the tube laminar, a reduction in resistance requires an increase in cross-sectional area, which in turn decreases pressure drop and requires that the pressure transducer used is more sensitive. The *Fleisch pneumotachograph*, is an example of a tubular flowmeter (Fig. 2.11) designed to have a very low resistance to respiratory flows and a very fast response rate. Its construction maximizes the pressure drop, ensures laminar flow measurement conditions with small-bore metal tubes, whilst minimizing the resistance by using a large number of short tubes. To prevent expiratory gas condensing in these tubes they are heated. However, a very sensitive and stable pressure transducer is required to make the respiratory flow measurements.

The operating principle of the *orifice flowmeter* is the resistance produced by a simple hole (Fig. 2.10B) that has been placed in a plate in the flow stream within the flowmeter. The orifice causes the flow downstream of itself to be turbulent whether the upstream flow is turbulent or not. The pressure drop achieved is thus proportional to the square of the volumetric flow rate, which makes calibrating an orifice flowmeter difficult, even over a limited range of flow rates. However, orifice flowmeters are cheap to make. Since the flow associated with the orifice flowmeter is essentially turbulent it seems an unlikely candidate for a transducer for respiratory reciprocating flows. However, wire screen respiratory flow heads, which effectively use multiple orifices in a similar way to the multiple tubes of the Fleisch pneumotachograph, have fast responses and low resistances to respiratory flows. The wire mesh in these flowmeters can be heated as in the Fleisch pneumotachograph to prevent condensation effecting the measurements.

The final type of differential pressure flowmeter is the *Venturi flowmeter* (Fig. 2.10C). In this type, the gas is forced through a smoothly narrowed portion of tube. The differential pressure is measured between points upstream of the Venturi and its 'throat' (narrowest point). Since the gas is compressed by the restriction it has to accelerate as it passes through the throat. This decreases the pressure in the throat relative to the upstream pressure. Pressure difference is approximately proportional to the square of the flow rate. Venturi flowmeters can be made very accurately but are expensive and are unsuitable for reciprocating respiratory flows.

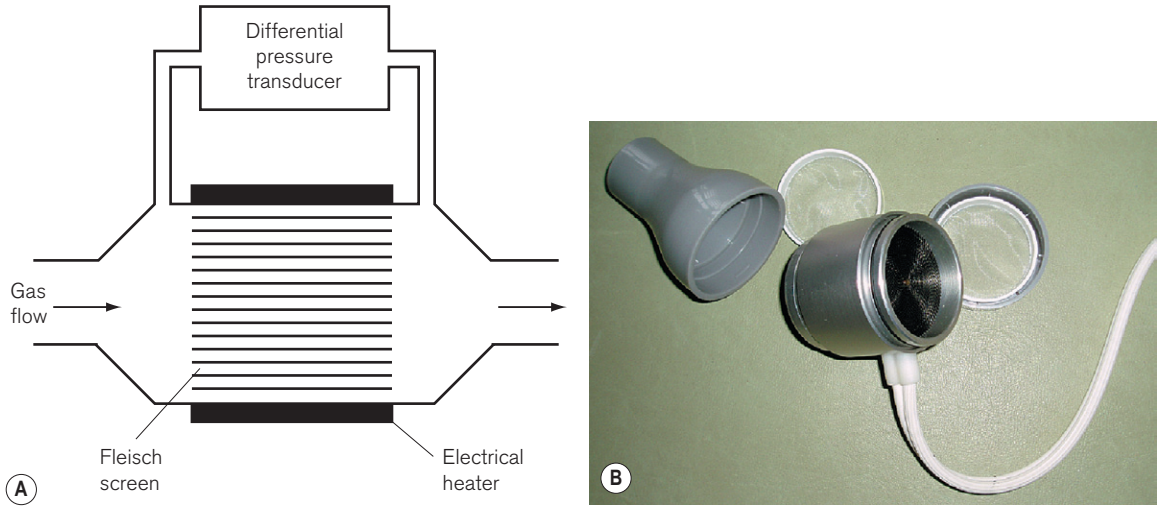


Figure 2.11 Fleisch pneumotachograph. **A.** Schematic and **B.** clinical example.

Variable-area constant differential pressure flowmeters

In the variable-area constant differential pressure flowmeter, the size of the orifice varies with changing volumetric flow rate to maintain a constant differential pressure. The most familiar design is known as a *rotameter.* Rotameter is actually a trade name of Elliot Automation, but the term is now ubiquitously applied to this type of flowmeter, regardless of detailed design or manufacturer. In the rotameter, a low mass bobbin is suspended by the gas flow in a transparent, vertically mounted (usually glass) tube with a tapering internal radius that is narrowest at its bottom.

During gas flow, an orifice is created by the annulus between the bobbin and the tube. At any given flow within the operating range of the rotameter, the bobbin will find a level at which the differential pressure created by this annulus results in a force upwards equal to the force of gravity downwards on the bobbin. As shown in Figure 2.12 for a given bobbin and tube design, this equilibrium point will occur at a certain point in the tube for a flow of, say 1.71 min^{-1}. If the flow is then increased to 61 min^{-1}, the bobbin will rise until the annulus is sufficiently large for equilibrium to be re-established. At low flow rates, flow becomes a function of viscosity because the comparatively longer and narrower annulus behaves like a tube. With higher flow rates, the annulus is shorter and wider and behaves like an orifice and is therefore density-dependent. Thus rotameters are only calibrated for a particular dry gas. The weight of the bobbin decides the pressure of operation of the rotameter and is thus constant.

The flow rate indicated by the rotameter is normally read from the top of the bobbin against a scale, but the exact reading point from the bobbin must be shown diagrammatically on the apparatus. Calibration of a rotameter can be calculated from physical principles by knowing the weight of the bobbin and the taper of the tube or, more practically, it can have its scale marked on to it when different constant flows of gas are applied. The scale is not normally marked from zero since, at flow rates giving positions near to the bottom of the tube, inaccuracies are more likely to occur. The minimum flow scale mark is the first with reliable accuracy.

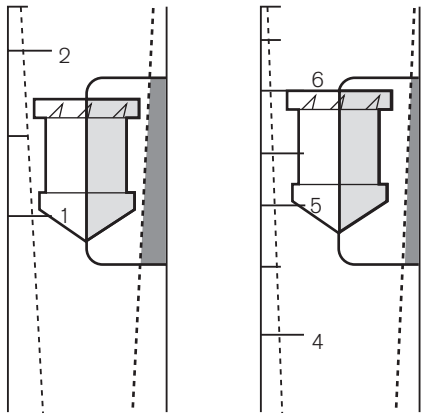


Figure 2.12 The rotameter. In each case, a portion of the tube has been cut away to show that the gap, or annulus, varies with the flow rate. The calibration should be read from the top of the float (e.g. in the right-hand diagram the flow rate is 6 l min^{-1}).

The scale on a rotameter need not be linear if a non-linear taper is used. Figure 2.13 shows a rotameter flow tube with a shallow taper for increased sensitivity at low flows, then a deeper taper for higher flow rates to allow

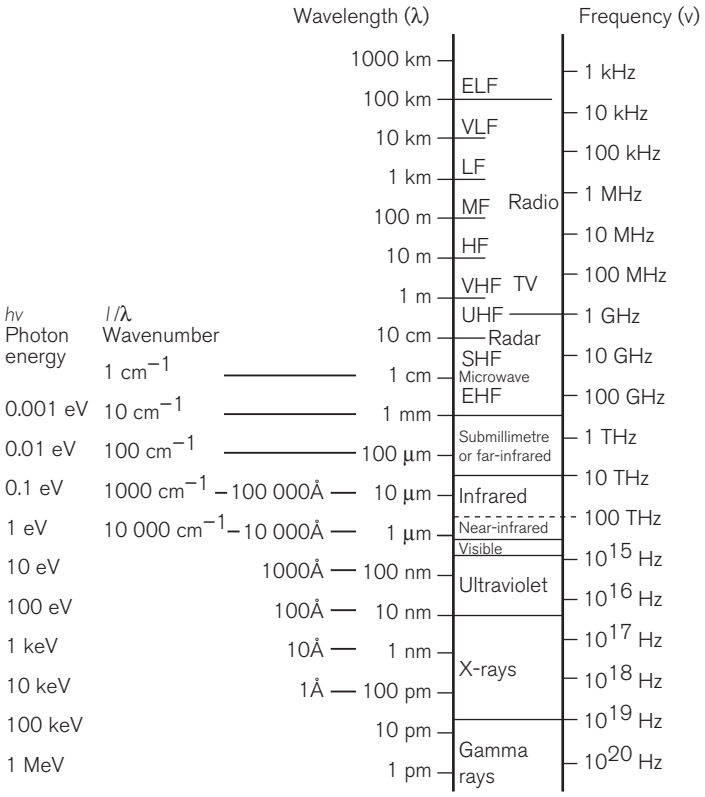


Figure 2.13 A flowmeter tube with varying taper to give an elongated scale at lower flow rates but allowing calibration for high flow rates also.

a greater range. An alternative arrangement, allowing improved accuracy, is the use of double flowmeter tubes in series (cascade flowmeters), where the first much narrower tube (and lighter bobbin) has a more gentle taper to allow an expanded scale at low flows (typically reading from 0.25 1 min^{-1} to 1.0 1 min^{-1}).

The accuracy of the rotameter is dependent on the bobbin being consistently in the centre of the tube. Variation away from the centre can lead to a change in effective annulus area or changes in flow pattern around the bobbin, affecting the consistency of the differential pressure generated. Thus rotameter accuracy is dependent on the tube being vertical. Most bobbins used in rotameters in anaesthetic equipment are also spin stabilized to keep them in the centre of the tube. This is done by cutting angled slots in the top flange of the bobbin (Fig. 2.12) to act as windmill vanes. A further design feature is to have a bobbin with a low centre of gravity so that it is more stable in the gas flow.

It is important that the only forces acting on the bobbin are those generated by the differential pressure due to flow and the force of gravity. Both static electricity and extra drag, due to dirt on the tube sides, can lead to underreading. Static electricity can be neutralized by having earthing wires or springs at the top or bottom of the tube. If the bobbin is not spinning easily, momentarily changing the flow to either zero or maximum, forces the bobbin to touch these wires and can restore normal working. However, in the case of wires at the top of the tube, it is possible in some older designs for the bobbin to become stuck and apparently indicate an erroneous high flow rate, so caution is recommended. Alternative precautions against the effect of static electricity can be to disperse the static on the tube by spraying it with water or coating the internal surface of the tube with an invisible thin conductive layer of gold.

Erroneous assumptions of zero flow can be made if the bobbin is not easily visible when at the top of its travel in the flowmeter tube. This risk has since been designed out by placing a stop lower down in the tube.

Constant temperature hot-wire anemometry

Until relatively recently, the only type of fresh gas supply flowmeter on the anaesthetic machine would have been the rotameter. However, the desire to integrate the functions of the anaesthetic machine with both patient

monitoring and ventilator control, in order to create a single integrated anaesthesia workstation, has produced the need to measure gas flows in ways that can be linked to electrical outputs. Differential pressure flowmeters, when combined with solid-state pressure transducers, can produce an electrical output of fresh gas flow and/or respiratory flows in ventilators. An alternative is the hot-wire anemometer (see below), which can now be used more freely than in the past as flammable anaesthetic agents have been taken out of service.

Anemometry is the measurement of the velocity of gas flow. If that flow is contained in a tube of fixed cross-sectional area, then it is easy to calibrate flow rate in terms of gas velocity. If a wire is heated to a fixed temperature above ambient and placed in a gas flow, the rate that heat is lost from the wire is dependent on the surface area of the wire, the density, viscosity and specific heat of the gas and the velocity of the gas flow across the wire. In fact, the rate of heat loss is proportional to the square root of the velocity of the gas.

In constant temperature hot-wire anemometry, the wire forms part of a Wheatstone bridge arrangement (Fig. 2.14), which measures the instantaneous resistance of the wire. The resistance will increase as cooling occurs. The output from the Wheatstone bridge is fed to an electronic amplifier, the output from which is used to control the current going to the hot-wire to keep it at a constant temperature/resistance. Thus, the output from the amplifier becomes a measurement of the heat transfer from the wire and hence the gas velocity.

Hot-wire anemometers (see Fig. 10.7) are very sensitive and have very fast response times. As well as being used as sensors of fresh gas flow, they can also be used as sensors of respiratory gas flows. However, since the heat transfer from the wire is dependent on the physical properties of the gas and hence its chemical composition, some care in calibrating the anemometer for changes in inspiratory/expiratory gas composition and humidity is required.

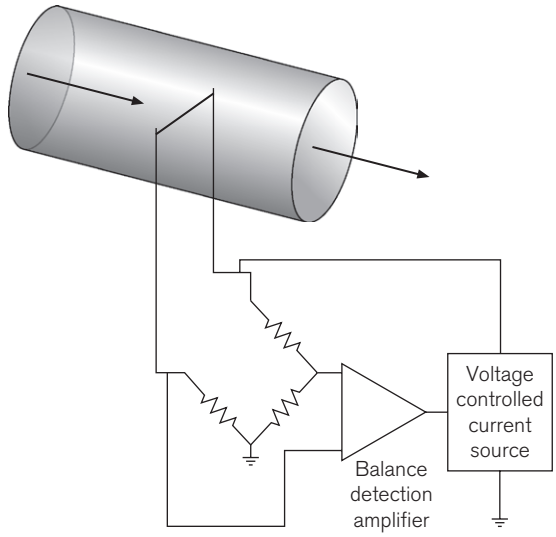


Figure 2.14 Schematic of a hot-wire anemometer in a Wheatstone bridge arrangement.

Mechanical flowmeters

The term 'mechanical flowmeter' can be conveniently adopted to cover all flowmeters that have a mechanism in the path of the flow that moves continuously at a speed that is proportional to the flow rate. In anaesthesia they have tended to be turbine-based.

The Wright's respirometer

The Wright's respirometer (Fig. 2.15A) is an inferential mechanical flowmeter, or rather integrated volume meter. It is inferential in that the turbine it uses does not rely on sensing the entire flow stream but allows some of the flow to leak past the turbine vane. Its principle is shown in Figure 2.15B.

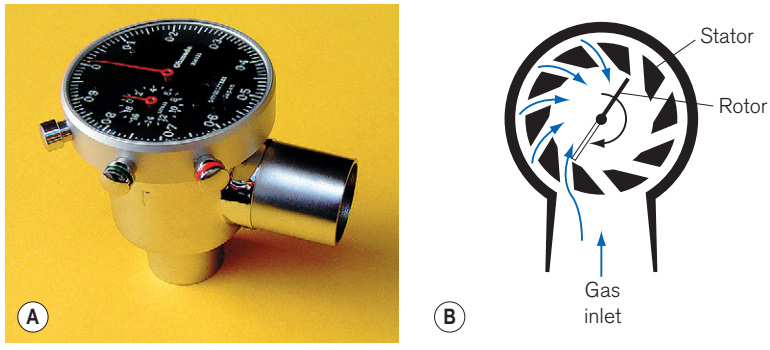


Figure 2.15 **A.** Inferential type flowmeter colloquially referred to as the Wright's respirometer. **B.** Plan view of the Wright's respirometer.

Gas flow to be measured is deflected through a tangentially slotted stator that directs the flow onto a turbine, giving the gas a circular motion (Fig. 2.15B). The turbine is of low mass and the gas flow imparts rotation, by giving up some of its momentum. At equilibrium, the rate of rotation of the turbine is proportional to the volume flow rate. Effectively, one rotation of the turbine is equivalent to the passage of a given volume of gas through the respirometer. Thus, it can be thought of as a volume integrator rather than a true flowmeter. In the original form, which is still on sale, the movement of the turbine is connected to a watch dial-like mechanism via a gearing mechanism and is calibrated in litres.

However, when used to measure exhaled gasses containing water vapour, condensation in the gearing mechanism leads to corrosion. Also the mechanical mechanism has a high inertia, which is added to the inherent inertia of the turbine and friction in the bearings of the turbine mounting. When following the flow curve of an expiration, this inertia leads to under-reading of integrated volume at low initial flows, followed by reasonably accurate measurement in the middle part of the breath, followed by over running in the late part of the breath as flow decreases. As a result, in normal adult breathing, the inaccuracies tend to balance out giving reasonably accurate measurements of tidal volume. However, at high and low breathing rates, the inertia effects can lead to poor precision.

To reduce inertia, the mechanical linkage has been replaced with a magnetic coupling of the turbine (Fig. 2.16) to an electronic scale, where a Hall effect transducer (see below) is used to count the number of rotations of the turbine. Hall effect transducers are semiconductor detectors that respond to very small changes in magnetic field.

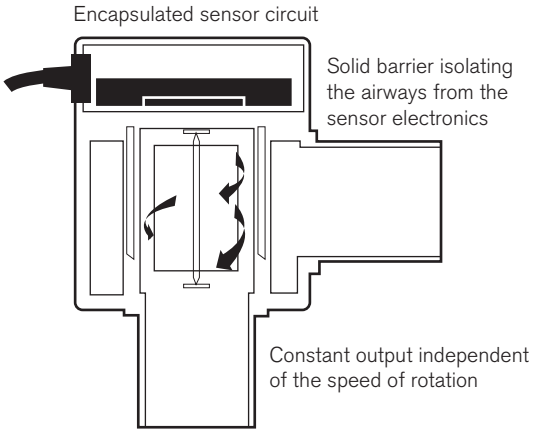


Figure 2.16 Side view of the electronic version of a Wright's respirometer.

The axial turbine flowmeter

In this system the turbine is mounted with its axis at the centre of the tube in which flow occurs. Upstream of the turbine there is a series of vanes which impart rotation to the flow. This rotation, in turn, imparts rotation to the turbine where, as in the Wright's respirometer, the rate of rotation of the turbine is proportional to the volume flow rate. Detection of the rotation is optical, with the turbine blades interrupting a light beam, which shines across the tube.

This design of flowmeter has been used in several guises, particularly in ventilators, over a number of years as a respiratory volume monitor. An example is illustrated in Figure 2.17. The advantage of the axial arrangement is that the turbine can be made lower mass and thus has less inertia than a vertical turbine, such as the Wright's respirometer. It thus suffers from fewer inertia inaccuracies and is accurate over a wider range of flow rates.

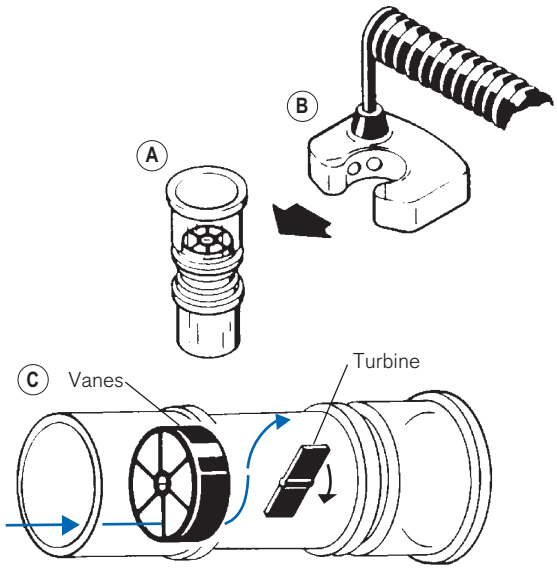


Figure 2.17 Sensor components for an axial turbine flowmeter used as a respiratory volume monitor. This sensor assembly is composed of two parts: **A.** cartridge; and **B.** optical sensor clip. **C.** Gas flow through cartridge.

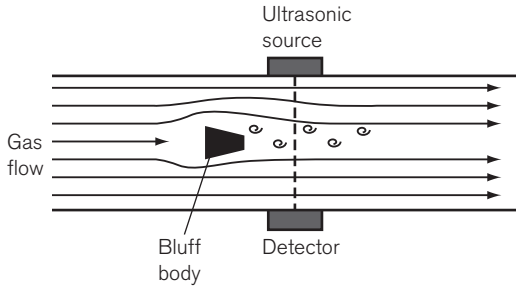


Figure 2.18 Vortex shedding flow transducer.

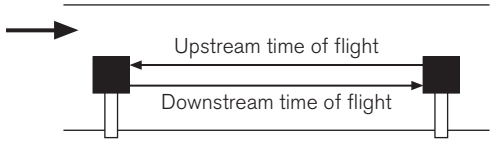


Figure 2.19 Time-of-flight transducer.

Ultrasonic flowmeters

There are two forms of ultrasonic flowmeter which have been used in anaesthesia: *vortex shedding* and *time of flight*.

The vortex shedding ultrasonic flow transducer

The schematic of a vortex shedding ultrasound flow transducer is shown in Figure 2.18. It consists of a tube in the middle of which is a bluff body that is designed to create vortices of turbulence downstream in its wake. The number of vortices that are shed is linearly proportional to the flow rate. The vortices are detected by their disruption of a narrow ultrasonic beam placed downstream at right-angles to the flow. By putting a second bluff body, facing the opposite way (to the right of the ultrasonic detectors in Fig. 2.18), the transducer can be made bi-directional and can detect reciprocating respiratory flows. Changes in gas composition can affect the rate of vortex shedding but the transducer is relatively insensitive to gas composition change.

The time-of-flight ultrasonic flow transducer

The schematic of the time-of-flight ultrasonic flow transducer is shown in Figure 2.19. Two ultrasound transmitter/receivers are placed, facing each other, in the middle of the flow. In an alternative design, they are placed facing at each other across the flow stream at an oblique angle. They alternately give ultrasound pulses that the other receives. The time-of-flight of these upstream and downstream pulses is measured. Upstream into the flow, the time-of-flight is increased by the flow velocity, whilst downstream it is reduced by the flow velocity. Given the distance between the transducers, the velocity of flow can be calculated from the length and the difference of the reciprocals of the two times of flight.

The measurement is actually a variety of anemometry. The absolute time-of-flight is dependent on the speed of sound in the gas and is therefore affected by gas composition changes.

Spirometers

Spirometer is a term used to describe a device that measures exhaled tidal volumes and flow state. Many anaesthetists will be familiar from their physiology practicals with the classic Tissot spirometer, where exhaled gasses are collected over water in an inverted bell. In anaesthesia practice, the most common spirometer is probably the Vitalograph single-breath spirometer used for preoperative assessment (Fig. 2.20).

Exhaled gas from the patient fills a lightweight bellows and causes a stylus to move across a chart. At the same time, the chart is moved at a constant speed at right-angles to the stylus movement. Thus, a chart of exhaled volume against time is plotted from which volumes and flow rates may be deduced.

Gradually this classical displaced gas design of spirometer is being replaced with electronic versions that use differential pressure flowmeters to measure exhaled flow. A microprocessor is then used to integrate the obtained flow to give the characteristics graphs of Fig. 2.20C, or to perform direct calculations of forced expiratory volume (FEV), forced expiratory volume in 1 s (FEV_1) and vital capacity (VC). This can be done in a hand-held device, such as the Vitalograph In2itive shown in Figure 2.21.

Peak flowmeters

Peak expiratory flowmeters are used for the surveillance and management of chronic obstructive airways disease. Anaesthetists come across measurements of peak expiratory flow (PEF) in the preoperative assessment of surgical patients and hence should be aware of some recent developments.

The Mini-Wright peak flowmeter (Fig. 2.22) has long been in use as a low-cost device for the day-to-day measurement of PEF. Several similar designs exist, all of which, in distinction to the Wright's respirometer (see above) and full size peak flowmeter, are of the variable outlet constant pressure type.

These devices carry their own inaccuracies and were calibrated originally according to the very first type of meter that was available. However, improvements in flow calibration methods, made possible by computerized pump systems, has led to concern over the last 10 years about the inaccuracies of these devices, particularly regarding over-reading in the mid-range of the meters. As a consequence, the European Community issued a new standard, EN 13826, enforced from 2004, which allows only devices meeting these more stringent accuracy requirements to be CE marked. Since 2004, Mini-Wright peak flowmeters have a new EU scale and clinicians need to become familiar with revised 'normal' values for PEF in health and disease states.

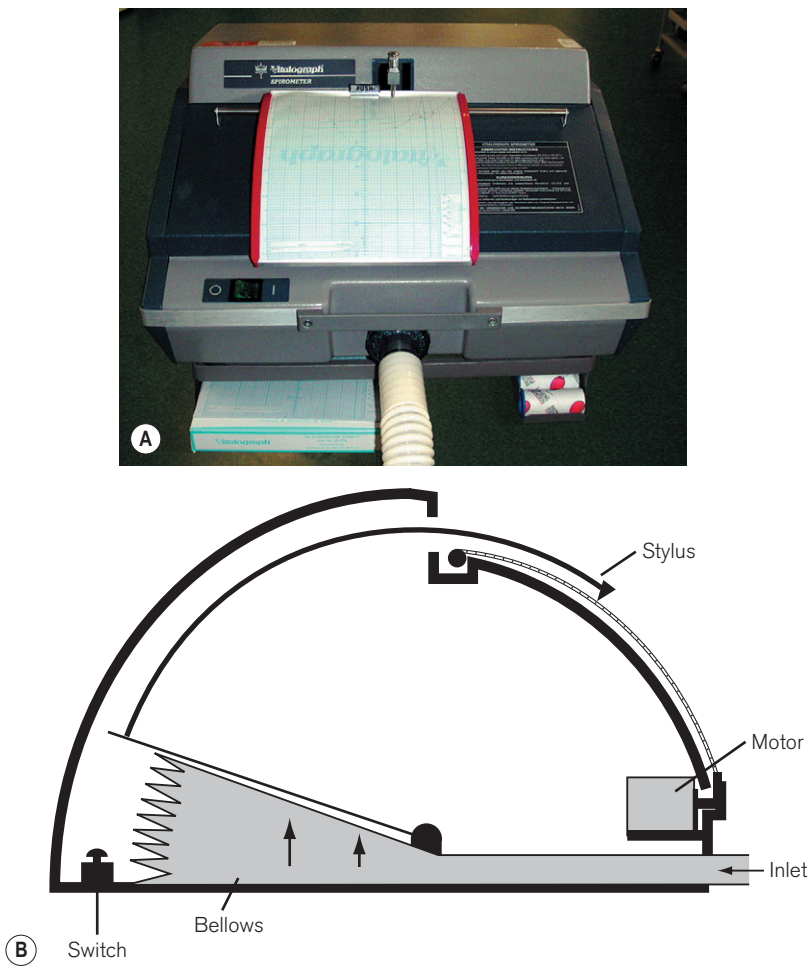


Figure 2.20 **A.** The Vitalograph bellows spirometer. **B.** Exhaled breath direct from the patient's mouth passes through the inlet causing the bellows to inflate and the stylus to mark the paper proportional to the exhaled volume.

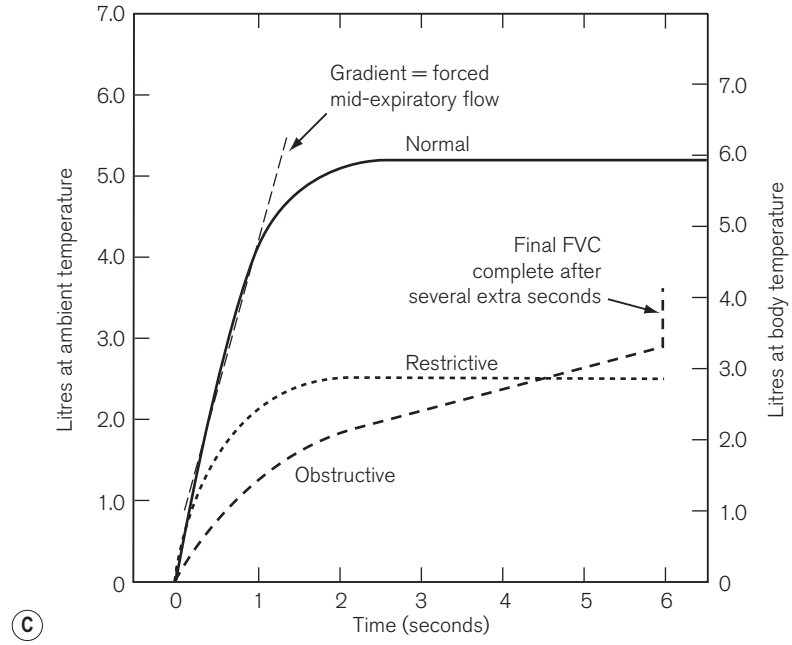


Figure 2.20, cont'd As the bellows lift off the switch, the paper carriage is moved horizontally by the motor at a constant rate so that exhaled volume is recorded against time. Typical tracings are shown in **C.**



Figure 2.21 Vitalograph In2itive handheld spirometer with Fleisch type pneumotacograph head.
Photo courtesy of Vitalograph Ltd UK.



Figure 2.22 The Mini-Wright peak flowmeter. Standard and low range versions with the different calibration scales (the long-standing Wright scale is now superseded in the EU, see text). From left to right: Wright-McKerrow (Wright) scale, European Union scale, American Thoracic Society scale.
Photo courtesy of Clement Clarke International Ltd.

FURTHER READING

Hemmings HC, Hopkins PM, editors. *Foundations of anesthesia: basic and clinical sciences*. 2nd ed. Edinburgh: Elsevier Science; 2004.

Miller MR. Peak expiratory flow meter scale changes: implications for patients and health professionals. *Airways J* 2004;**2**:80–2.

Roberts F. Measurement of volume and flow in gasses. In: *Anaesthesia and intensive care medicine*. Kidlington, UK: Medicine Publishing Company; 2003.

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Chapter 3

Vaporizers

Andrew J Davey

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Many inhalational anaesthetic agents are liquids under normal storage conditions and need to be in a vapour form before they can be administered to a patient. In order that they may be administered safely, an understanding of the phenomenon of vaporization is required.

LAWS OF VAPORIZATION

Molecules of a liquid have a mutual attraction for each other (a phenomenon called cohesion), which is sufficiently great for them to remain in close proximity. But they also possess varying degrees of kinetic energy and are in constant motion, colliding with each other. If the liquid has a surface exposed to air or other gasses, or to a vacuum, some molecules with a high kinetic energy will escape from this surface, resulting in the process of evaporation or vaporization. The molecules from the liquid, which exist in the gaseous phase, are known collectively as a vapour. This vapour exerts a pressure on its surroundings, which is referred to as vapour pressure. If the space above the liquid is enclosed, some of the molecules that have escaped while moving freely in the gaseous state will collide with the surface of the liquid and re-enter it. Eventually, there will occur an equilibrium in which the number of molecules re-entering the liquid equals the number leaving it. At this stage the vapour pressure is at a maximum for the temperature of the liquid and so is called the saturated vapour pressure (SVP).

Factors affecting vaporization of a liquid

Temperature

Vaporization is increased if the temperature of the liquid is raised, since more molecules will have been given sufficient kinetic energy to escape. Fig. 3.1 shows the vapour pressure curves of volatile anaesthetic agents (as well as water) and shows how they vary with temperature. If the liquid is heated, a point is reached at which vaporization now occurs not only at the surface of the liquid, but also in vapour bubbles that develop within its substance. The liquid is now boiling and this temperature is its boiling point. At this temperature, the SVP of the liquid is equal to the ambient atmospheric pressure.

The boiling point of a liquid may therefore vary with atmospheric pressure. At high altitudes (where the air is thinner, has a lower ambient pressure and therefore exerts less pressure on the surface of a liquid) there is a significant depression of the boiling point. This may render the administration of agents with low boiling points, such as ether, difficult. Fig. 3.2 shows the depression of the boiling point for water with change in atmospheric pressure.

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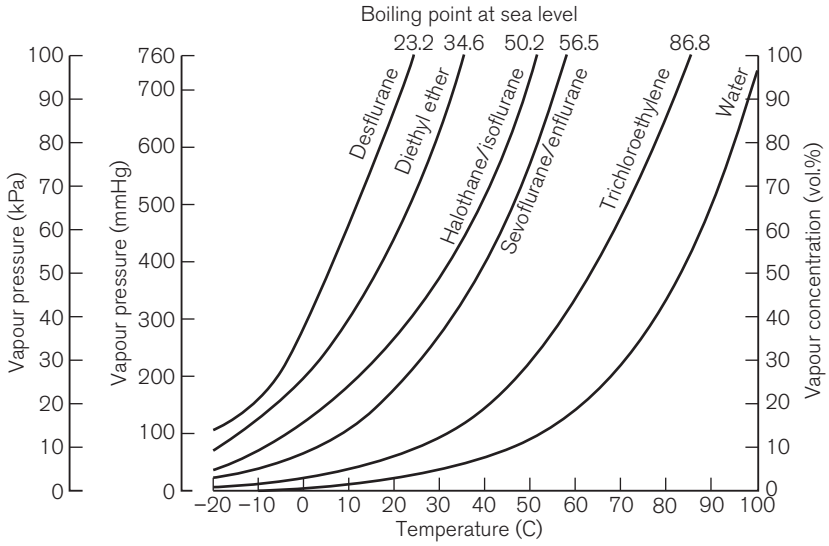


Figure 3.1 Vapour pressure curves for anaesthetic agents.

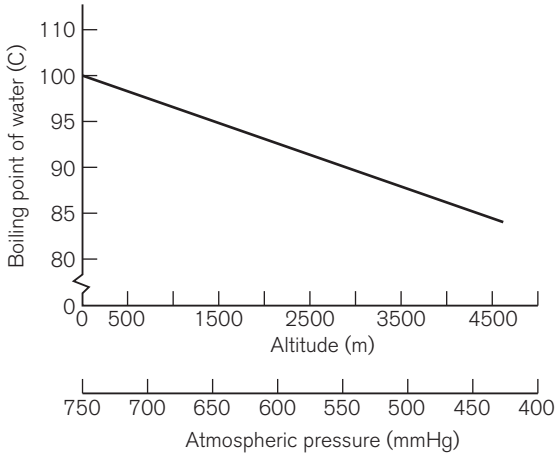


Figure 3.2 Variation of boiling point of water with atmospheric pressure or altitude.

Volatility

The speed at which a liquid vaporizes depends not only on its ambient temperature and pressure, but also on its volatility. A more volatile liquid has weaker cohesive forces between its molecules, such that they require less energy (i.e. a lower temperature) to vaporize (Fig. 3.1). This is reflected in a higher SVP at any temperature and a lower boiling point.

The surface area of the liquid

The greater the surface area of the liquid, the more space there is for molecules to leave the liquid. The rate of vaporization is, therefore, also proportional to the surface area of the liquid.

Removal of vapour from the vicinity of liquid

If the container holding the liquid is not closed, molecules will still leave the liquid and some will escape into the atmosphere. Some, however, will collide with adjacent vapour molecules and be bounced back into the liquid. If a gas is passed across the surface of the liquid, vapour will be removed more quickly allowing fresh vapour to form. Vaporization is, therefore, proportional to gas flow (convection) across the surface of the liquid (provided the temperature of the latter remains constant).

A liquid at a given temperature has a mixture of molecules with varying energies. Vapour molecules entering the vapour phase tend to be the ones with the highest energy (the hottest). The remaining liquid molecules have a lower average kinetic energy (and, therefore, a lower temperature). Fewer molecules remain with sufficient energy to form a vapour and so vaporization decreases.

VAPORIZING SYSTEMS

The various organic liquids that possess anaesthetic properties are too potent to be used as pure vapours and so are diluted in a carrier gas such as air and/or oxygen, or, nitrous oxide and oxygen. The device that allows vaporization of the liquid anaesthetic agent and its subsequent admixture with a carrier gas for administration to a patient is called a *vaporizer*.

Variable bypass vaporizers

Design features

Surface area of contact between carrier gas and the liquid

Vaporizers, which are required to be very accurate, should always present a saturated vapour to the bypass gas across a wide range of flows. To ensure sufficient vaporization at the highest planned flow, a sufficiently large surface area of liquid should be present. This size is also governed by the volatility of the agent used. A highly volatile liquid will require a smaller surface area. The surface area for vaporization of a liquid can be increased by causing it to spread (by capillarity) over a large sheet of porous material which may be folded in such a way that the carrier gas passes across its entire surface (Fig. 3.5).

Temperature

As vaporization progresses, the vaporizing liquid as well as the vaporizer cools and the quantity of vapour produced decreases. In an attempt to retain the performance of the device, the temperature drop is minimized or prevented by the incorporation of a heat source (heat sink).

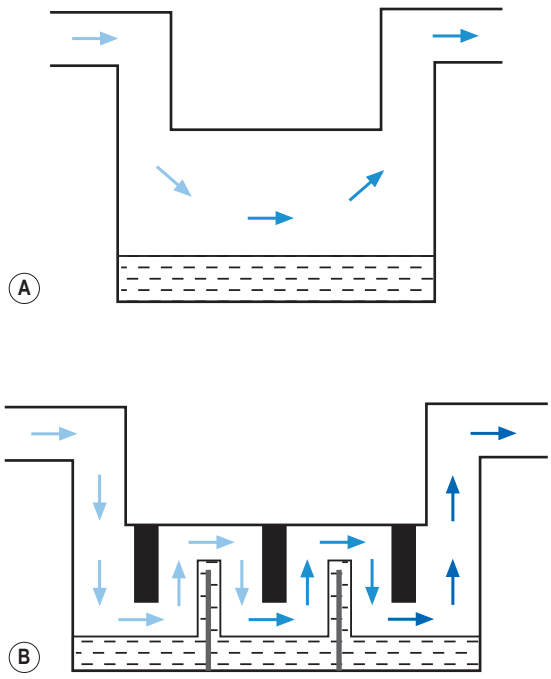


Figure 3.5 Vapour pick-up in vaporizers. **A.** Carrier gas passing over a small surface area of liquid anaesthetic agent. **B.** The surface area for vaporization is increased by porous wicks dipped into the liquid. The carrier gas is also made to pass close to the surface of the liquid by the baffles, thus increasing vapour pick-up.

This normally takes the form of a water bath or substantial metal jacket or even a heating element surrounding the vaporizing liquid. Metal jackets and water baths, however, can only transfer a finite quantity of heat and so only minimize the inevitable fall in temperature.

In order to maintain the expected output of the vaporizer when this occurs, a greater proportion of carrier gas is required to pass through the vaporizing chamber in order to collect sufficient vapour molecules. This is achieved by using devices that are sensitive to changes in temperature (*temperature-compensating devices*, Fig. 3.6) and which then increase the flow through the vaporizing chamber.

Two types are commonly used:

1. The first (Figs 3.6A and B) consists of two dissimilar metals or alloys placed back to back (i.e. a bi-metallic strip). As the two metals have different rates of expansion and contraction with temperature, the device has the ability to 'bend'. It can, therefore, be used to vary the degree of occlusion in the aperture of a gas channel (usually the bypass) and thus alter the flow of carrier gasses through it.
2. In the second arrangement (Figs 3.6C and D), the bi-metallic device consists of a central rod made of Invar, a metal alloy with a low coefficient of expansion, sitting inside a brass jacket, the top part of which is attached to the roof of the vaporizing chamber. The rod is attached only at the base of the brass jacket, which has a higher coefficient of expansion. The outer surface of the jacket is immersed in liquid anaesthetic agent in the vaporizing chamber. As the aforementioned liquid cools, the brass jacket contracts more than the Invar, which is pushed upwards into the bypass, restricting the flow of bypass gas and increasing the flow of carrier gas through the vaporizing chamber.

In heated vaporizers, the heating elements are thermostatically controlled. They are, therefore, automatically temperature compensated and do not require the addition of the devices above.

Potency of anaesthetic agent

As described above, current anaesthetic vapours are too potent to be administered as saturated vapours and require suitable dilution. Therefore, only a proportion of the gas intended for the patient is diverted in the vaporizer to collect vapour. This amount may be varied to produce the desired concentration by using an adjustable flow-splitting valve (see Fig. 3.3). This is usually a rotary valve incorporated within the vaporizer outlet. It proportions the flow of gas between the vaporizing chamber and the vaporizer bypass system, thus controlling the final vapour composition (i.e. the more gas going through the vaporizer chamber, the greater the amount of vapour leaving the vaporizer). The flow-splitting valve is calibrated in percentage of the vapour in the final gas/vapour composition.

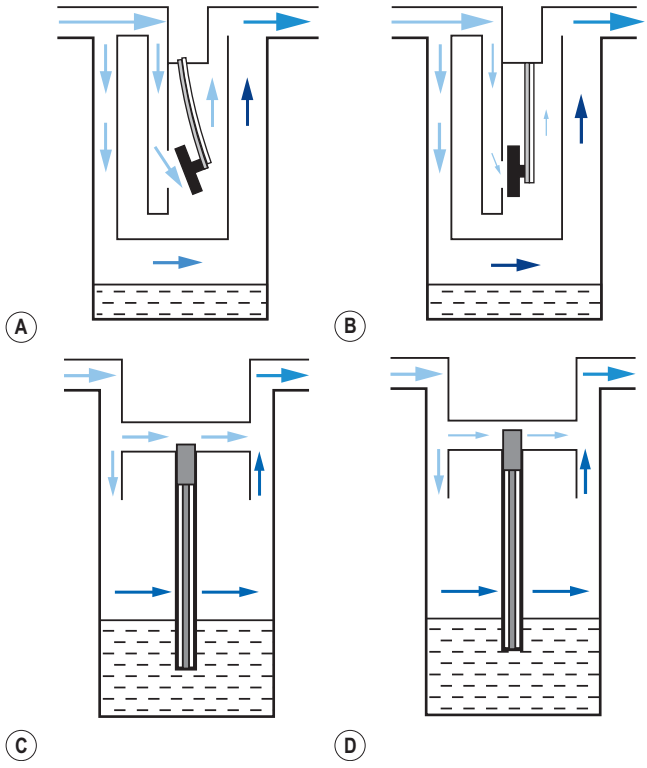


Figure 3.6 Temperature-compensating devices. **A.** The bi-metallic strip in a vaporizer bypass operating at ambient temperature. **B.** The same device operating at a cooler temperature; the bi-metallic strip has moved closer to the inflow increasing the resistance in the bypass and increasing the amount of gas passing through the vaporizing chamber and therefore increasing vapour pick-up. **C.** A bi-metallic arrangement that works on a similar principle. The inner rod is made of Invar, a relatively non-expansile metal. The outer jacket is in contact with the vaporizing liquid and is made of an expansile metal (brass). **D.** When this contracts (with cooling) it drags the choke on the inner rod into the bypass, increasing the resistance to flow through it.

However, this valve is accurate only if the vaporizer is temperature-compensated (see above). As both the temperature-compensating mechanism and the flow-splitting valve work by altering resistance through the vaporizer, the devices are dependent on each other. Therefore, each vaporizer for a designated anaesthetic agent is individually calibrated at the factory (see below) for that agent and for a specific range of temperatures and flow rates of carrier gas.

As the potency of anaesthetic agents varies widely, the flow-splitting ratios must be individually tailored for each agent and vaporizer design. Agents with high potency will require a wide splitting ratio so that a smaller amount of gas passes through the vaporizer. This produces a lower final concentration when mixed with bypass gas.

Volatility

The flow-splitting ratio is dependant on the volatility of the agent. For example, at any given temperature, a very volatile agent will produce a higher saturated vapour pressure than a less volatile agent, even though they may have similar potencies. The former, however, requires a flow-splitting valve with a wider ratio so as to increase the dilution in order to provide a similar concentration. Table 3.1 shows the relative potency and volatility of some liquid anaesthetic agents that influence vaporizer design.

Types of variable bypass vaporizers

Draw-over vaporizers

The early vaporizers relied on the patient's respiratory effort to draw gas over the vaporizing surface (hence their name). Unfortunately, draw-over systems are subjected to very variable flow rates, i.e. from 0 to 601 min^{-1} (the peak inspiratory flow in a hyperventilating adult). At these higher flows the carrier gas may fail to pick up a saturated vapour resulting in a reduced concentration leaving the vaporizer. Furthermore, the gas pathways must offer little resistance to flow so as not to compromise the patient's inspiratory effort. This restricts the design of the vaporizer

Table 3.1 Relative potency and volatility of some liquid anaesthetic agents

	VOLATILITY			POTENCY
Agent	**Boiling point (°C) at 100 kPa**	**SVP at 20°C (kPa)**	**SVP at 20°C (mmHg)**	**MAC (vol.%) in 100% O_2**
Desflurane	23	88.5	669	6
Diethyl ether	35	57.9	440	19
Isoflurane	48	31.5	240	1.15
Halothane	51	31.9	243	0.76
Enflurane	56	23.1	175	1.68
Sevoflurane	58	21.3	156–170	2

MAC, minimum alveolar concentration;
SVP, saturated vapour pressure

components, especially the flow-splitting valve (see above), which must have sufficiently wide a bore. It is very difficult to design a flow-splitting valve that will work accurately over a wide range of flow rates, i.e. 1–60 l min^{-1}. As discussed above, the valve must present a low flow resistance so that at flows of 40 l min^{-1} (the peak flow in a patient breathing spontaneously), no respiratory embarrassment is caused. However, if the flow across this valve drops to about 4 l min^{-1}, the resistance through the valve will be so low that carrier gas will preferentially pass across the bypass channel rather than through the vaporizing chamber where it has to mix with and then push the 'heavy' vapour out into the attached breathing system. At this flow and below, there is thus bound to be a marked fall in vaporizer performance.

Plenum vaporizers

A vaporizer can be made more accurate if the carrier gas is pressurized to make it as dense as the vapour, so that at lower flows it more readily mixes with this rather than tending to pass above it in the vaporizing chamber. Furthermore, if a smaller, continuous flow of gas, i.e. 0–15 l min^{-1}, is used, there is a less rapid removal of vapour, ensuring that a saturated vapour is present at all times. This allows the vaporizer to be calibrated very accurately. This type is usually referred to as a plenum vaporizer (plenum being the term which describes a pressurized chamber). The typical flow resistance (2 kPa (20 cm H_2O) at 5 l min^{-1}) found in plenum vaporizers renders them unsuitable for use as draw-over vaporizers. With plenum vaporizers, the problem of high intermittent flow rates in a breathing system, generated by a spontaneously breathing patient or a mechanical ventilator, is accommodated by siting a reservoir (bag or bellows) downstream, which stores inspiratory gas and vapour during the exhalation phase.

However, flowmeters on an anaesthetic machine are calibrated for use at or around atmospheric pressure. Therefore, the final design of a plenum vaporizer develops from a compromise between the high carrier gas pressures required for accurate vapour delivery and the low pressures required to maintain the accuracy of the flowmeters.

FACTORS AFFECTING VAPORIZER PERFORMANCE

Extremes of temperature

It is obvious that a temperature-compensating mechanism can operate only within a reasonable temperature range. At too low a temperature, vaporization will be low, and it may be uncontrollably high when it is too hot.

Barometric pressure

Ideally, a vaporizer should also be calibrated at a specific barometric pressure. Strictly speaking, as a saturated vapour is only altered by temperature, one might expect the calibration of a vaporizer to be independent of barometric pressure. However, changes in barometric pressure will affect the carrier gas composition passing through the vaporizer, which in turn will affect the concentration of vapour in the mixture leaving it. For example, when the barometric pressure is reduced (at altitude), the number of molecules of carrier gas flowing through the vaporizer is reduced. However, the number of vapour molecules collected by the gas in the vaporizing chamber remains unchanged, although these now represent a higher percentage of the total number of molecules leaving the vaporizer. The final output concentration of vapour is thus

now higher than at sea level, although its partial pressure remains unchanged. As anaesthetic effect is governed by the partial pressure of the agent in the body, there is no effective change under the normal conditions. However, extremes of pressure may have a significant effect. For example, at very high altitude (low barometric pressure), a very volatile liquid such as ether may boil at ambient temperature. This may render the use of such agents difficult. Fig. 3.2 shows the variation of boiling point with atmospheric pressure.

Pumping effect

When a resistance is applied to the outlet of the anaesthetic machine, such as that which occurs when manually assisted inspiration or controlled ventilation is used, there is an intermittent increase in the anaesthetic gas pressure, which is transmitted back to the vaporizer. When this happens, it causes carrier gas within the vaporizer to be compressed. Gas in the outlet is already saturated and, therefore, cannot pick up any more vapour. When the back pressure is released, the expanding carrier gas, which is also saturated with vapour, surges out through both the inlet and outlet of the vaporizer chamber. The gas that leaves the inlet enters the bypass and adds to the vaporizer output to increase in the final vapour output. (Fig. 3.7 demonstrates the sequence of events.)

This effect can be minimized by the fitting of internal compensating mechanisms. It may be achieved by either:

- increasing the resistance to flow through the vaporizer and bypass so that the carrier gas develops a higher pressure within the vaporizer, so as to reduce the pumping effect. However, the pressure increase due to vaporizer design should be as small as possible as these pressures are transmitted back to the flowmeters, which are calibrated for use at near atmospheric pressure; or
- building an elongated flow passage into either the inlet or outlet of the vaporizer to minimize the effect of surges in pressure (Fig. 3.8).

Some vaporizer designs employ both mechanisms. The former cannot be fitted to draw-over vaporizers as they would produce too great a resistance to flow (see below, Temperature-compensated vaporizers).

Furthermore, where plenum vaporizers are fitted, some anaesthetic machines now incorporate a non-return valve on the end of the back bar, so that the back pressure surges on the vaporizer are reduced. However, pressure still builds up to some extent in the back bar when the non-return valve closes due to higher downstream pressure.

Liquid levels

The liquid level within the vaporizing chamber may affect performance. If the vaporizer is overfilled, insufficient

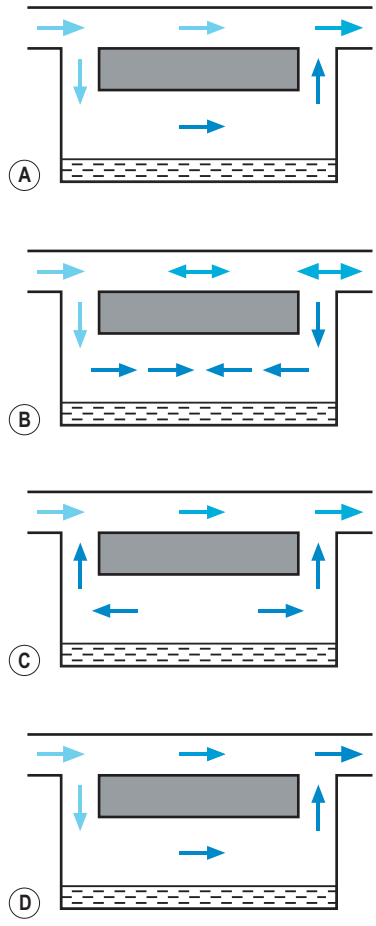


Figure 3.7 **A.** Flow-through vaporizer with no back pressure. **B.** Back-up pressure and/or reverse flow causing build-up of carrier gas in the vaporizer chamber. **C.** Release of back pressure causing gas and saturated vapour to escape through the vaporizer inlet and outlet and into the bypass. **D.** Anaesthetic vapour in the bypass gas added to that from the vapour outlet so increasing the final vapour concentration.

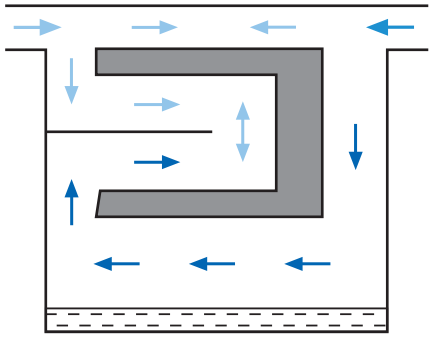


Figure 3.8 Elongation of the inflow channel in a vaporizer preventing saturated vapour reaching the bypass.

exposed surface area of wick may cause a drop in vapour output. Additionally, over-filling may result in dangerously high concentrations, due to spillage of liquid agent into the bypass.

Anaesthetic agents

The anaesthetic agent halothane contains a stabilizing agent, thymol. This is a waxy substance which, if left in the vaporizer, would clog the felt or cotton wick found in older models, reducing the potential surface area for vaporization. This would then reduce the vaporizer performance. Thymol may also 'gum up' the vaporizer, making the control knob difficult to adjust, as well as compromising the internal mechanism. The manufacturers, therefore, used to advise that the liquid agent be drained off and replenished at intervals. This advice should be tempered by consideration of economy and the frequency with which the vaporizer is employed. Recent models from most manufacturers for use with halothane have wicks made from synthetic materials that do not absorb thymol into the fabric, so many can now recommend a service interval (often with a change of wick) of 5 years. Vaporizers used with other agents often have a 10-year service interval.

Anaesthetic agents are susceptible to various types of degradation including decomposition in sunlight. Hence, all these agents are stored in dark brown glass or treated metal containers. Sevoflurane is also degradable by *Lewis acids* (such as metal oxides and metal halides) to hydrofluoric acid and other toxic compounds. Water inhibits such degradation. Hence, different manufacturers add varying amounts of water to the compound for stabilization. For example, Abbott incorporates 352 ppm water whereas other generic versions contain 19–57 ppm. Some low water formulation agents have been shown to undergo substantial degradation by reacting with the internal metal components of some vaporizers. This damage and the accumulation of toxic by-products present a potential patient safety issue. Care must taken to identify those vaporizers at risk and use only high water content sevoflurane in them. Some manufacturers coat the insides of their vaporizers with Teflon in order to be able to use the generic products.

Carrier gas composition

Vaporizer output may be affected when the carrier gas composition is changed. This is due to changes in viscosity and density, which alter the performance of the flow-splitting valve. Increasing the concentration of nitrous oxide reduces the vapour concentration. This is of little importance in clinical practice at present. However, the interest in the gas xenon (which is five times as heavy as air), as a potential anaesthetic, may change this. There is also a further mechanism causing a change in vaporizer output when nitrous oxide concentrations are increased. Nitrous oxide dissolves in volatile agents, so that the effective total gas flow through the vaporizer is temporarily reduced.

Stability

Some vaporizers, if tilted or inverted, may allow the liquid agent to contaminate the bypass. This has caused a fatality in the past when a vaporizer was accidentally overturned prior to attachment to the machine. Modern designs of vaporizer aim to eliminate this risk by sealing off the vaporizing chamber from the bypass and outflow gas channels when the vaporizer is set to *off* or *transport mode* as they must be to allow disconnection from the back bar of the anaesthetic machine.

Summary of vaporizer performance

Vaporizer performance can thus be affected by:

- Temperature (unless the vaporizer includes some compensatory device that minimizes the effect of temperature, such as a heat sink and/or temperature compensator)
- Flow. All vaporizers are affected to some degree by flow (performance data are usually available from the manufacturer). Plenum vaporizers perform better than draw-over vaporizers
- Barometric pressure (minimal clinical effect in practice)
- Variable vaporizer working pressures (back pressure surges)
- Liquid levels within the vaporizer
- Movement and tilting of vaporizers (see below, under Specific vaporizers)
- Carrier gas composition
- Stabilizers in the inhalational agent (e.g. thymol).

CALIBRATION OF VAPORIZERS

Vaporizers designed to give an accurate output are individually calibrated prior to leaving the factory. Typically, they are filled with the designated anaesthetic agent and left in a room at a standard temperature (23°C) for 4 hours. A blank control dial (linked to a computer) is attached and rotated at various carrier gas flow rates. The output concentration is measured using a sample that is analyzed by a refractometer (see Fig. 15.3). The dial (which has a unique serial number) is then removed and a calibration scale etched onto it from the information stored on the computer. It is then re-attached to that same vaporizer and the calibration confirmed prior to leaving the factory. Vaporizers may also have the calibration confirmed in a similar manner, following servicing.

FILLING OF VAPORIZERS

In the original plenum vaporizers, a screw-threaded stopper in the filling port was simply unscrewed, liquid agent was added and the stopper reconnected. These systems were criticized and have largely disappeared from use as the vaporizer could easily be filled with the wrong agent. Despite this, they are still supplied by all the major manufacturers for certain countries (Fig. 3.9) and are referred to as 'screw fill' systems.

Agent-specific filling devices (in the UK) (Fraser Sweatman pin safety system) were introduced by Cyprane (which became Ohmeda and is now part of GE Healthcare) in the early 1980s. In these the distal end was keyed to fit a vaporizer calibrated, labelled and colour coded for a specific agent and the proximal end keyed to fit only the neck of the bottle for that agent (Fig. 3.10A).

This is now referred to commonly as the 'key fill' system. Dräger have a similar system (Dräger-fil) for their Vapor range (Fig. 3.10B) as do GE Healthcare (Easy-Fil) (Fig. 3.10C).

Although these devices go some way to reducing the potential for filling vaporizers with the wrong agent, it is by no means foolproof. For example it would still be possible



Figure 3.9 Screw fill port on a vaporizer.



Figure 3.10 **A.** Fraser Sweatman pin safety system for isoflurane. Note that the top has unique grooves (shape, colour and position) that fit only the specific filling ports on matching vaporizers. **B.** The Dräger-Fil system for isoflurane. **C.** GE Healthcare Easy-Fil for generic sevoflurane.



Figure 3.11 **A.** A sealed bottle of Abbott sevoflurane with its integral uniquely shaped filler (Quick-Fil). Note that the filler on the bottle is secured by a crimped metal seal. **B.** Similarly, a sealed bottle of desflurane; with Saf-T-Fil system. The filler caps are shown alongside the bottles. As the contents are pressurized at ambient temperature, the glass bottle is encased in a plastic coat to prevent it exploding if damaged.

to decant one agent from an open bottle to another agent-specific one. Some manufacturers of sevoflurane (Abbott) and desflurane (Baxter) produce sealed bottles to which the agent-specific filling device is already fitted and made tamper-proof with a crimped metal seal. These filling devices also have valves, which are only opened when inserted fully into their respective filler ports, so as to prevent spillage (Figs 3.11A and B). The filling system for sevoflurane and the older agents can be designated at ordering of the vaporizer but sevoflurane from Abott Laboratories only uses their own, Quik-Fil system (Fig. 3.11A). The desflurane system (Fig. 3.11B) is called Saf-T-Fil (Baxter Healthcare) and operates across all vaporizer manufacturers.

Although it is increasingly difficult to fill a vaporizer with the wrong drug, it is far easier to fail to fill a vaporizer because of the multitude of filling systems between drug and vaporizer manufacturers.

EXAMPLES OF VARIABLE BYPASS VAPORIZERS

Temperature-compensated vaporizers

TEC 5 (GE Healthcare)

Fig. 3.12 shows a TEC 5 (with a keyed filler), its mode of operation and performance charts. Although now owned by GE Healthcare, these vaporizers and some TEC 7s (see below) are still badged as Datex–Ohmeda.

When the vaporizer is attached to the back bar of an anaesthetic workstation, gasses destined for the patient pass through it via three channels. In the OFF position, with the rotary valve (5) closed, the gasses pass through the inlet (1) and into one channel (6). From here the gasses pass across the top of the vaporizer, without coming into contact with the vaporizing chamber and leave through the outflow (9).

In the ON position, this pathway closes and the other two channels (2 and 7) are open. The vaporizer channel has an elongated passage (2) that funnels carrier gas into the centre of the wick assembly. This comprises a hollow tube of Teflon cloth held open by a steel wire spiral and wound into a helix (3) within the vaporizer. The lowest helix is in contact with the liquid agent. This arrangement greatly increases the surface area for vaporization over previous Tec models. The gas passing through the wicks becomes saturated with vapour and enters a rotary valve assembly (calibrated control knob).

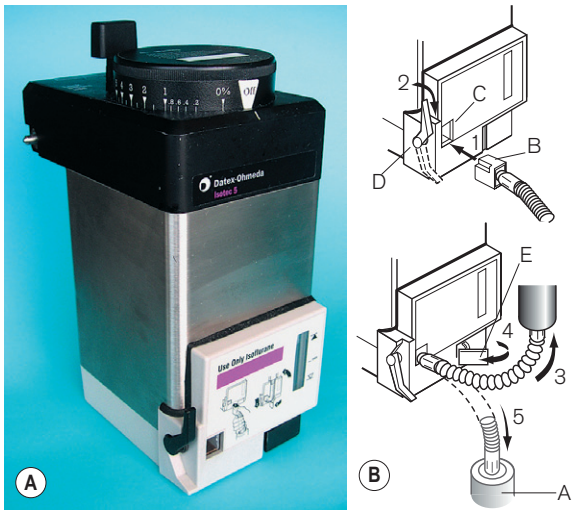
The second channel guides bypass gas through the base of the vaporizer, through the temperature compensating device and towards the outflow (9).

The proportion of gasses passing the two channels is determined by (a) the resistance to flow in the temperature-compensating device (8), and (b) the calibrated control knob (5) which varies the resistance from the vapour chamber exit (6). The percentage of vapour at the outlet depends on the amount of vapour-laden gasses that is mixed with the fresh gasses passing through the bypass. As the temperature within the vaporization chamber falls (reducing the vapour concentration produced), the thermostatic valve closes. This causes a greater proportion of the total gas flow to pass through the chamber; by this means the vapour concentration in the output is kept constant.

The vaporizer has some significant design features. If it is accidentally inverted, the liquid agent will not spill into the bypass. It also incorporates an interlock facility. If two vaporizers with this latter feature are placed on a back bar (see below), the first vaporizer to be switched on extends lateral rods that impinge on the adjacent vaporizer preventing its operation. Also, the vaporizer dial cannot be turned if the vaporizer is improperly mounted on the anaesthetic machine, i.e. not seated correctly and locked on to the back bar.

TEC 7 series

These vaporizers (Fig. 3.13B) are identical in function to the TEC 5. However, they also offer a wider selection of filler assemblies and a change in the cosmetic appearance over the TEC 5 (Fig. 3.12) to match the manufacturer's workstation.



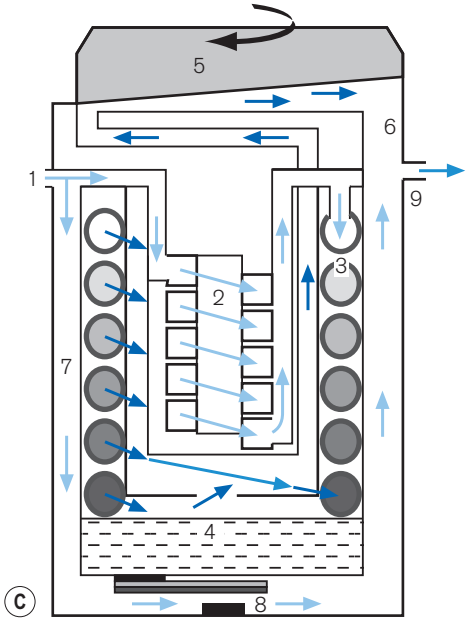


Figure 3.12 **A.** The TEC 5 vaporizer. **B.** Filling the vaporizer. (1) insertion of keyed filler B into vaporizer filling port C, (2) applying the lock D to make the filler/vaporizer connection secure, (3) inverting and raising the bottle of agent to create a filling pressure, (4) opening the chamber lock E to fill the chamber, (5) lowering the bottle below the vaporizer to empty the chamber if required. **C.** Working principles of Tec 5 and Tec 7 vaporizers. (1) inlet, (2) elongated passage that prevents 'pumping effect', (3) helical wick, (4) base of vaporizing chamber, (5) rotary valve for metering vapour saturated carrier gas, (6) mixing chamber, (7) bypass, (8) temperature-compensating device, (9) outlet.



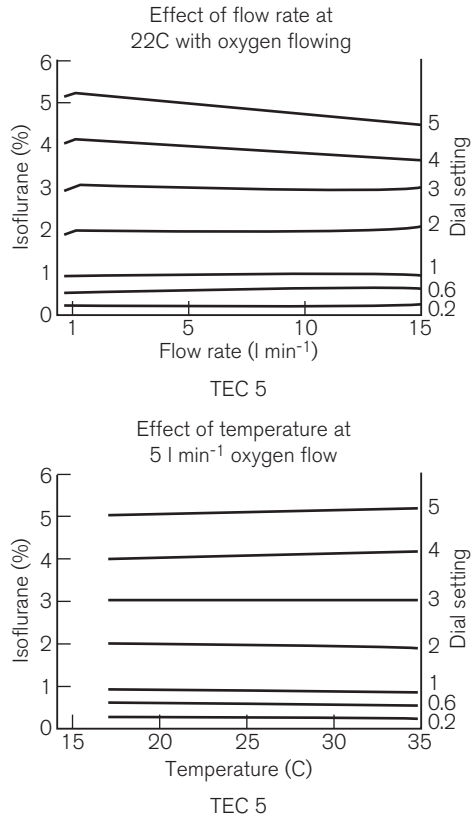


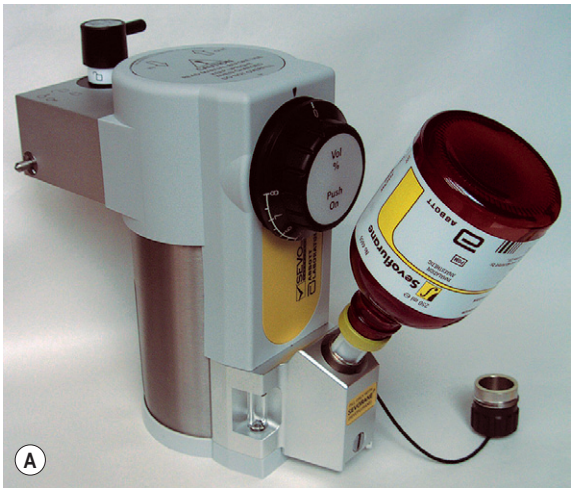
Figure 3.13 **A.** TEC 7 isoflurane and sevoflurane vaporizers. **B.** Performance of TEC 5 and TEC 7, as influenced by changes in flow rate and temperature.

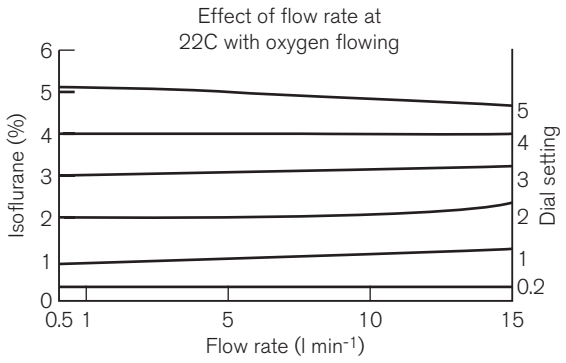
Blease Datum

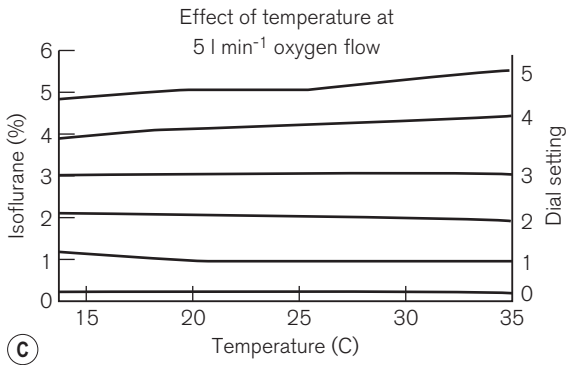
Blease is now part of Spacelabs Healthcare and the vaporizer range is now labelled as Spacelabs Healthcare. However, the design is still identical to the latest Blease Datum and there are many still labelled as such in circulation.

This range fulfils all the criteria for a temperature compensated vaporizer. It has:

- a heat sink
- a thermal compensating device
- a large surface area for vaporization
- an elongated inlet to minimize the damping effect.







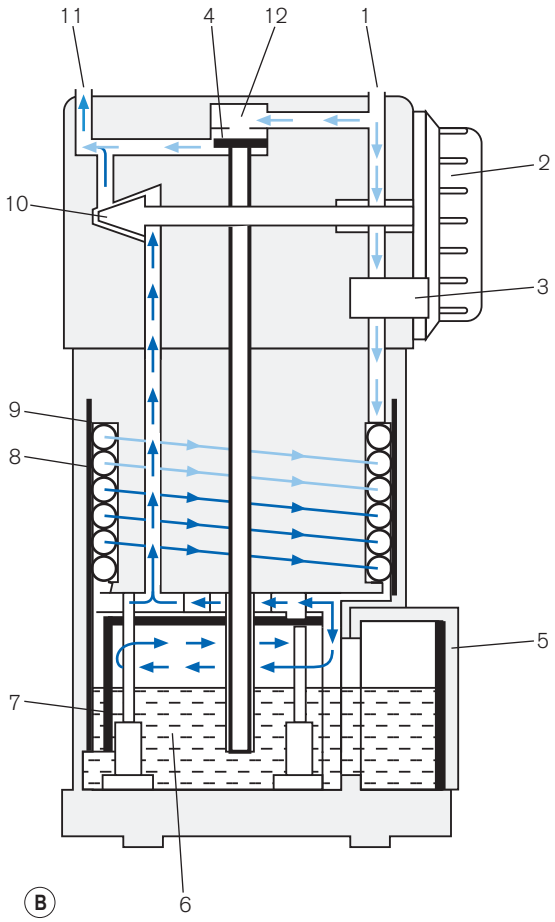


Figure 3.14 **A.** The Blease Datum vaporizer with agent-specific filler for sevoflurane. **B.** Working principles: (1) fresh gas input, (2) control dial, (3) zero lock, (4) thermal compensator rod, (5) filler block for agent-specific device, (6) agent reservoir, (7 and 9) wick extension, (8) elongated Teflon wick, (10) vapour control valve, (11) combined gas and vapour output, (12) variable resistance bypass valve. **C.** Performance characteristics for variations in temperature and flow.

The earlier version had a very large heat sink made of brass, in order to reduce the cooling effect of vaporization: hence the vaporizer weighed 11 kg. The latest version (Fig. 3.14A) has an improved thermal conductivity and now weighs 7.5 kg. The earlier version had a stainless steel outer covering but this was separated from the brass body of the vaporizer by a layer of plastic. The latter has been removed to improve the absorption of radiant heat from the surroundings and has enabled less brass to be used. The heavy weight of the vaporizer helps to seat the vaporizer firmly on the back bar when it is connected to the anaesthetic workstation and so reduces the potential for leaks. The working principles of the vaporizer are shown in Figure 3.14B.

The wick is made from a long tube of PTFE (Teflon) wadding that is supported internally by a wire coil (8). This is wound into a spiral to create a large surface area for contact with carrier gas that passes through the tube. At the base of the spiral, the Teflon is made to drape into the liquid reservoir (7). The material has a high capillarity to ensure that the wick is saturated at all times, even when the reservoir is low (liquid level compensation).

In the OFF position, there is a 'zero lock' (3) on the dial that isolates the vaporization chamber so that all the patient designated gas travels through the bypass. In the ON position this gas is split into two flows. One passes through the elongated wick (8), collecting vapour; the elongated passage behaves as a damping device to counteract the pumping effect. From here, the gas, which is now saturated with anaesthetic vapour, travels through to the vapour control valve (10), operated by the control dial (2). It then joins the remainder of the gas in the bypass. A thermal compensator (4) alters the flow through the bypass (12), to correct for changes in vapour production at lower temperatures. The bi-metallic device consists of a central rod made of Invar, a metal alloy with a low coefficient of expansion, part of which sits inside a brass jacket, the top part of which is attached to the roof of the vaporizing chamber. (see above: Temperature compensating devices: p. 44, and Fig. 3.6C and D).

Unlike the TEC vaporizers, the control dial may be turned on when it is not connected to a machine and so if this is left on and the vaporizer is inadvertently tipped, there is the potential for liquid anaesthetic agent to enter the bypass. However, the relevant channels are placed towards the front of the vaporization chamber near the filler block. If the vaporizer falls on its side, the filler block prevents these channels from being submerged in liquid agent. Figure 3.14C shows the performance curves for the vaporizer.

Dräger 'Vapor' 2000 series of vaporizers

Figures 3.15A, B and C show the vaporizer, working principles and performance curves. Models in the range are all compensated for temperature and pumping effects. In the OFF position, gasses destined for the patient are directed through the bypass (12) in the vaporizer without coming into contact with anaesthetic vapour.

When the vaporizer control dial is switched ON, these gasses split into two flows with one part initially flowing through a series of baffles (2) that counteract pressure surges (the pumping effect). From here, it is directed through the vaporizing system, where it becomes saturated with vapour. This takes the form of a tubular wick (3) coiled in a spiral, through which the gas passes. The outer surface of the spiral is attached to a sleeve of similar material (4) that dips into the liquid anaesthetic agent in the reservoir (5), so as to keep the spiral wick soaked. The wick is made of a synthetic material with a high capillarity, but which does not absorb agent. Therefore, a stabilizing agent, such as thymol that is added to halothane, will not clog the wick and reduce its efficiency. This allows the vaporizer to be used for prolonged periods between services.

From the vaporizing chamber, the saturated gasses pass to a conical control valve (6), whose aperture is adjusted by the calibration dial (7). From here they pass to a mixing chamber (9), where they blend with bypass gasses prior to leaving the vaporizer. If the operating temperature of the latter drops, a compensating device (8) (see also Figs 3.6C and D) proportionately decreases the flow of gasses through the bypass so as to maintain the output of the vaporizer.

In the OFF position, the vaporizing chamber has a small connection to atmosphere (11) that allows some gas to escape when liquid agent is added. This makes the filling process easier.

Penlon Sigma Delta vaporizer

The Penlon Sigma Delta, shown in Figs 3.16A and B, includes all the features of a modern Plenum vaporizer.

In the OFF position, gasses destined for the patient are directed via the inlet (1) through the bypass (2) to the outlet (3). A closing mechanism (not shown) prevents any gasses coming into contact with anaesthetic vapour in the vaporizer.

When the control knob (4) is turned on, the closing mechanism is released and a second channel is opened that ducts a portion of these gasses through the vaporizing system. They pass initially through a helical damping coil (5) that prevents saturated vapour tracking back through the vaporizer and contaminating other gasses in the back bar (the 'pumping effect').

From here they pass into the vaporizing chamber (6) and around the wick (7). The latter is novel in that it is made of sintered polyethylene (1 m long), which is held in close proximity to a copper backing plate. The two are then coiled into a spiral, the top and bottom of which are made gas-tight. The carrier gasses, therefore, have to

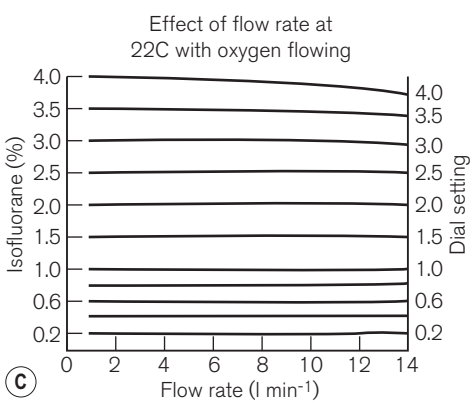
pass around the spiral, coming into contact with the whole surface area of the wick. The wick assembly is designed as a cartridge for ease of removal and cleaning, and has a long service life. The recommended service interval is 5 years for halothane vaporizers and 10 years for all others.

Gasses saturated with anaesthetic vapour leave the chamber and pass through an orifice whose aperture is varied by a needle valve (8) attached to the control knob. The latter, therefore, controls the amount of vapour-laden gasses flowing through the device to mix with the bypass gasses.

A bi-metallic temperature-compensating element (TCE) (9) is placed in the vaporizing chamber, so that its base is immersed in the vaporizing liquid. When the liquid agent cools, the aperture of the TCE closes and diverts more gas through the vapour chamber pathway, in order to maintain the accuracy of the vaporizer output.

The vaporizer is interesting in that it is relatively light (5 kg), being made of aluminium. Aluminium has a better thermal conductivity than brass and so, despite its weight, it conducts a similar amount of radiant heat as does an equivalent sized brass device. The rotary control can be turned on when the vaporizer is disconnected from an





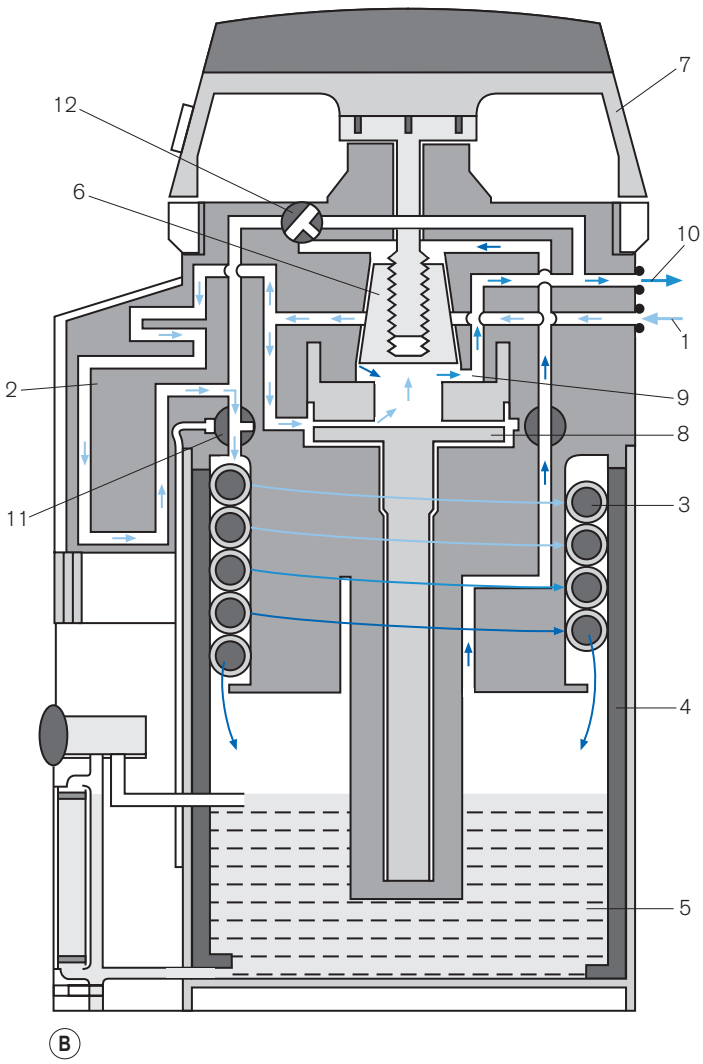
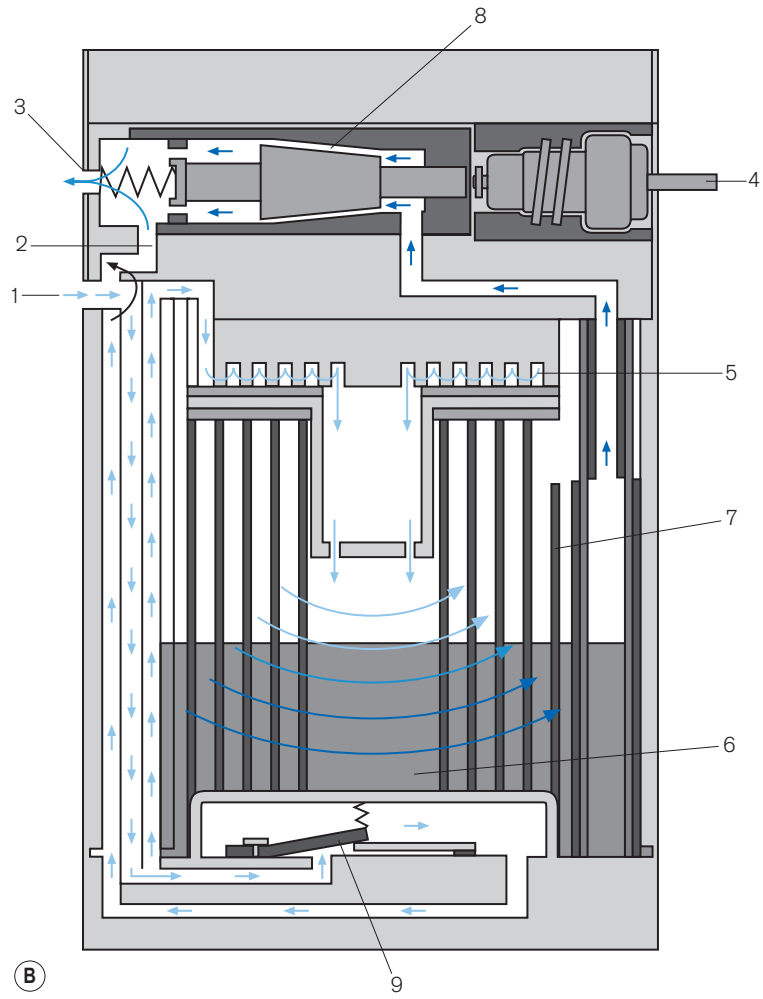
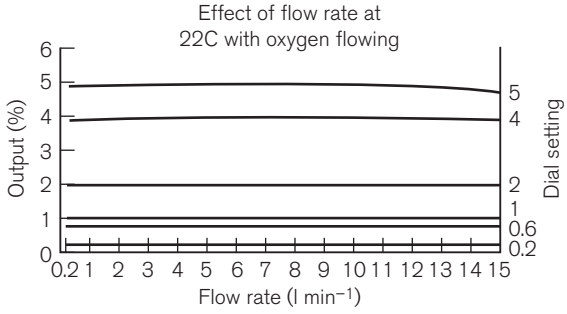


Figure 3.15 **A.** Dräger Vapor 2000 vaporizer. **B.** Working principles: (1) gas input, (2) damping the chamber, (3) tubular wick, (4) wick extension, (5) reservoir chamber, (6) concentration control valve, (7) calibration dial, (8) temperature compensation device, (9) mixing a chamber, (10) output, (11) bypass tap, (12) bypass tap. **C.** Performance characteristics.







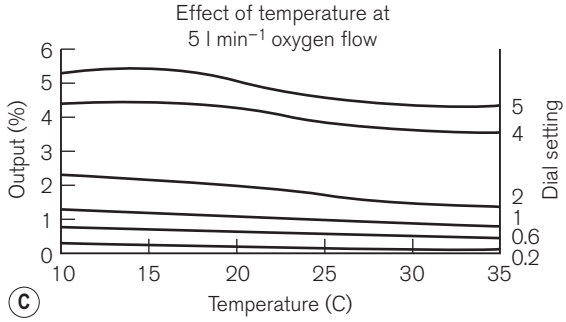


Figure 3.16 **A.** Penlon Sigma Delta vaporizer. **B.** Working principles: (1) inlet, (2) bypass, (3) outlet, (4) control knob, (5) helical damping coil, (6) vaporizing chamber, (7) wick, (8) needle valve, (9) temperature compensating element. **C.** Performance characteristics.

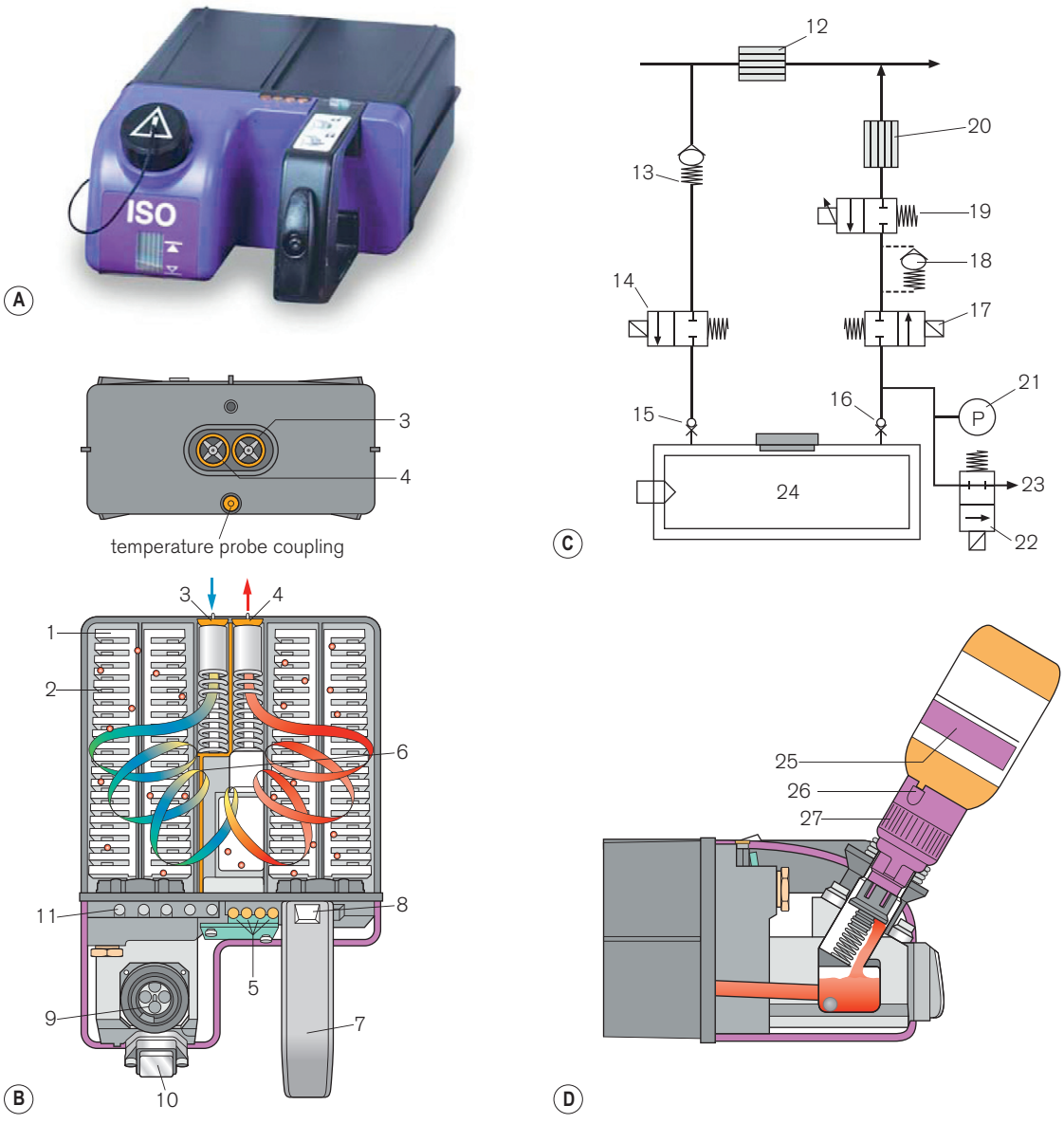


Figure 3.17 **(A)** Aladin 2 cassett. **(B)** Diagram of the back of the casette and cutaway showing carrier gas flow – (1) wicks, (2) lamellae, (3) inlet valve, (4) outlet valve, (5) contact for electronic temperature sensor and liquid anaesthetic agent level, (6) temperature sensor strip (7) handle, (8) locking mechanism, (9) fill system (10) liquid level window, (11) agent identification magnets (not visible externally). **(C)** Diagram of control unit- (12) bypass flow measurement, (13) one way valve, (14) inflow close valve, (15) inlet check valve, (16) outlet check valve, (17) outflow close valve, (18) liquid flow prevention, (19) proportional flow valve, (20) agent flow measurement, (21) cassette pressure sensor, (22) cassette pressure relief valve, (23) flow to scavenging, (24) Aladin cassette. **(D)** Filling the casette: (25) Colour coded bottle, (26) Colour coded adapter tab, (27) Colour coded adapter cap.

with an electronic pressure sensor (21). If the outflow pressure exceeds 2.5 bar, a relief valve opens (22) and vents the flow to scavenging (23).

Draw-over vaporizers

All the plenum vaporizers described above offer resistance to the gas flow. For this reason carrier gasses have to be pressurized to some degree to be driven through them. However, pressurized gas sources are not always available in some countries or in certain situations. Draw-over vaporizers, with their low-resistance gas pathways, can be installed within a breathing system and are therefore, a useful alternative to plenum systems despite not being as accurate. Fig. 3.18 illustrates various breathing systems in which a draw-over vaporizer has been installed. In systems A–D, exhaled gasses are vented to the atmosphere and suitably scavenged where appropriate. However, in system E, the patient's exhaled gasses are recirculated through the vaporizer. This is of importance, since not only will

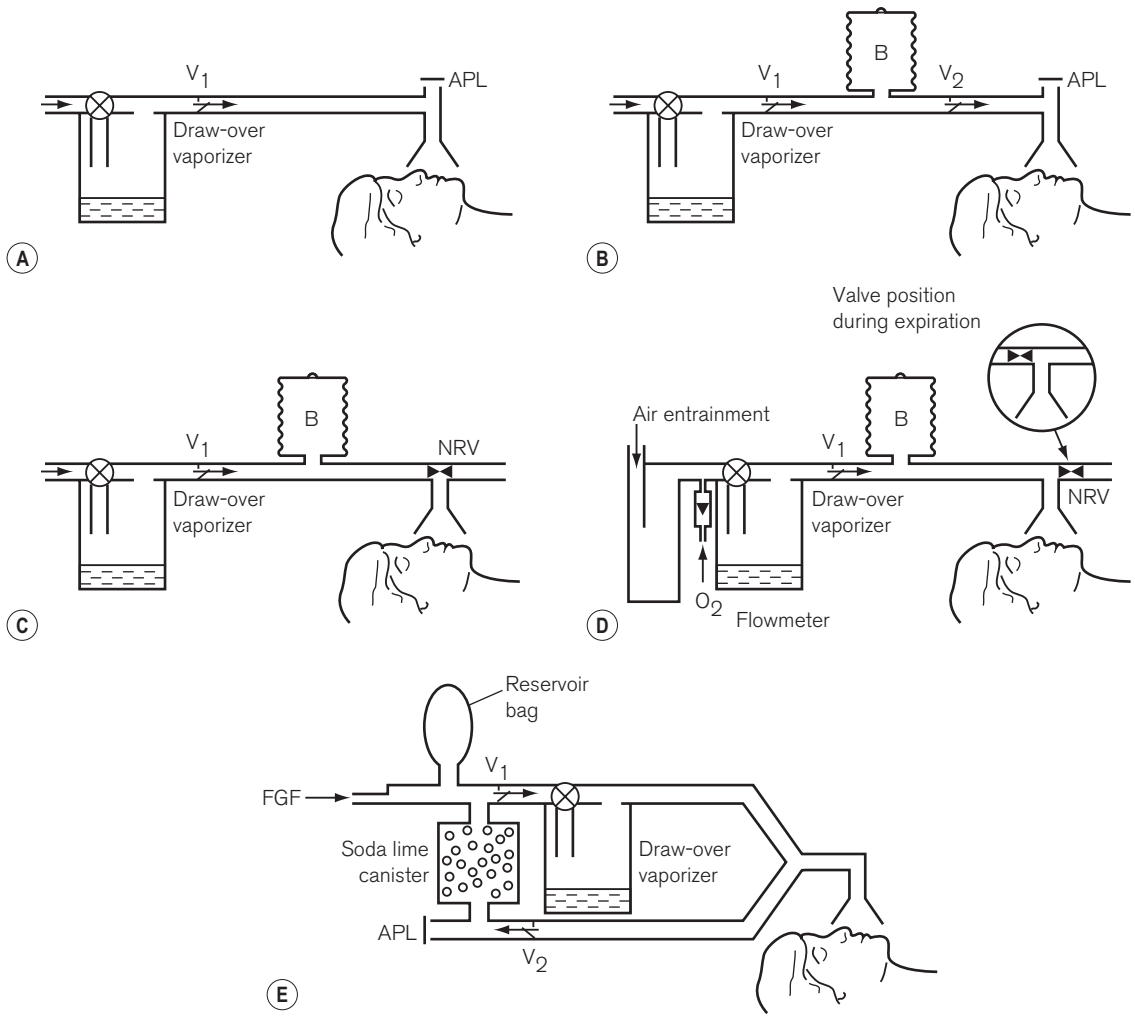


Figure 3.18 Draw-over anaesthetic systems. Note that they all contain non-return valves to prevent reverse flow through the vaporizer. System **A.** is used for spontaneous respiration. System **B.** incorporates a bellows and therefore requires a second non-return valve (V_2). If this is used for controlled ventilation (system **C.**), an additional non-return valve (NRV) is often substituted for the APL valve. If the former is of a design which has a tendency to jam, the second valve V_2 is either removed, or in the case of the Oxford Inflating Bellows, held open by a magnet. In system **D.** an oxygen flowmeter has been added. During the expiratory phase, the continuing supply of oxygen flows into the reservoir and is stored for use in subsequent breaths. In system **E.** the vaporizer has been placed in a circle breathing system. A vaporizer in this position is often referred to as a vaporizer in circle (VIC).

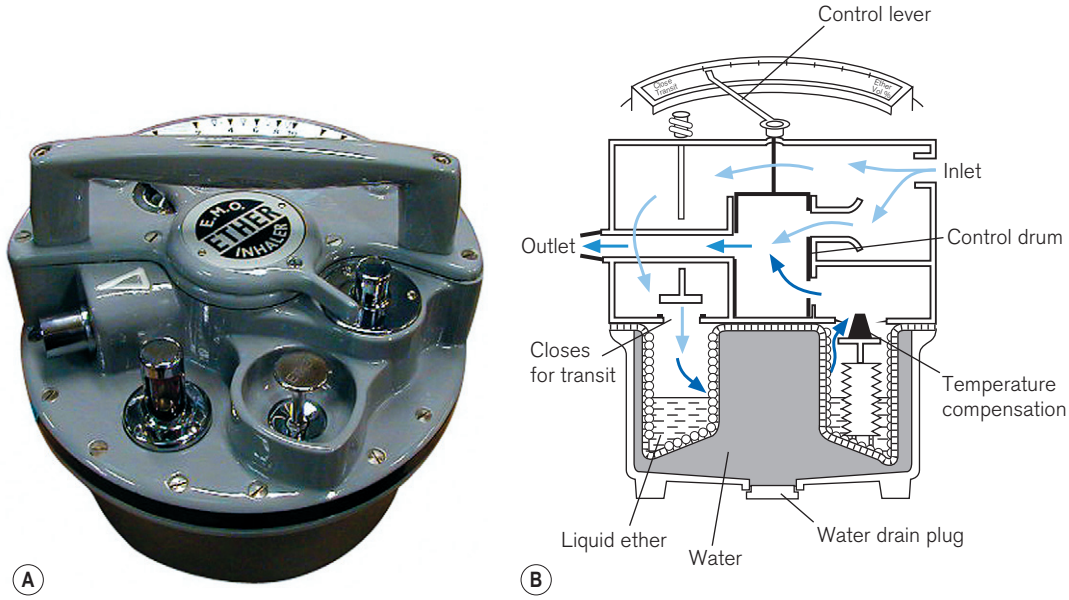


Figure 3.20 **A.** The EMO ether inhaler. This is a low-resistance vaporizer which is both temperature and level compensated. **B.** Working principles. Note that there is a mass of water, which provides a heat sink. When the control lever is put to the 'close for transit' position, the ether chamber is sealed off to prevent spillage.

Epstein, Macintosh, Oxford (EMO) ether inhaler

This has been deservedly the most popular draw-over vaporizer (Fig. 3.20) for the administration of ether, and is still widely used throughout the world. For spontaneous respiration, it is often used in conjunction with the OMV (Oxford Miniature Vaporizer). The latter is usually filled with halothane to provide smooth and rapid induction of anaesthesia, which is then continued by ether from the EMO. Both vaporizers may be used in conjunction with self-inflating bellows for techniques employing controlled ventilation.

MEASURED FLOW VAPORIZERS

TEC 6 (Plus) (Desflurane)

This vaporizer (Figs 3.21A and B) was designed specifically for the volatile agent desflurane. This agent is unusual in that its boiling point is around room temperature and so it would not remain as a liquid in the reservoir of a conventional vaporizer. It therefore requires an unusual design which dispenses with most of the conventional compensating devices mentioned above.

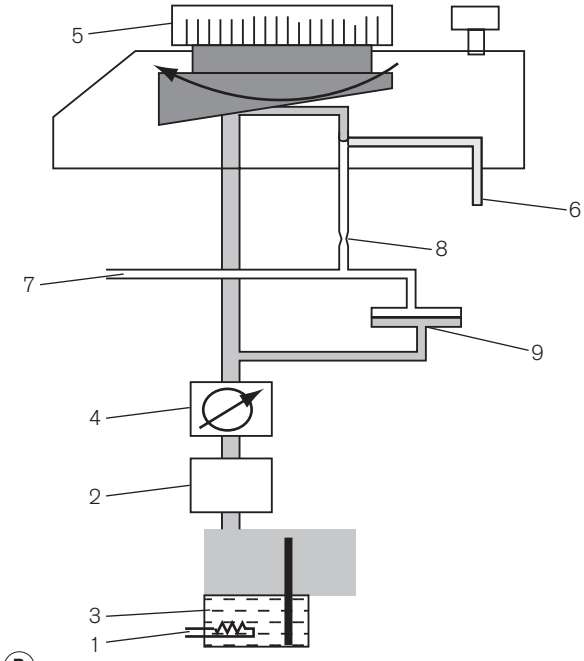
The reservoir of the TEC 6 has two thermostatically controlled electric heating elements (1), which raise the temperature of the desflurane to 39°C. At this temperature, the SVP is 194 kPa (1500 mmHg). When vapour is required, a shut-off valve (2) opens and pure vapour under pressure is allowed to escape from the reservoir (3). It passes to an electronic pressure regulator (4) that reduces the pressure to that normally found in a plenum vaporizer (1–2.5 kPa) and then to a variable flow restrictor linked to the calibrated concentration selection dial (5), from where it is fed into the carrier gas flow leaving the vaporizer (6).

Fresh gas flow into the vaporizer (7) has to pass through a narrow restriction (8) so that its pressure (which increases with flow) matches that normally found in a plenum vaporizer. With increasing flows, two independent sensors (9) in the pathway detect the pressure rise and instruct the desflurane pressure regulator to increase proportionately the desflurane pressure (and flow) to the selection dial, so as to maintain the set vapour concentration. If the readings from the two sensors are not similar, the shut-off valve closes and isolates the vaporizing chamber.

The vaporizer has a number of other features:

- The vaporizer heaters are switched on automatically when the unit is connected to the electricity supply. However, a 5–10 min warm-up time is required to reach operating temperature. During this time the concentration dial cannot be turned on.





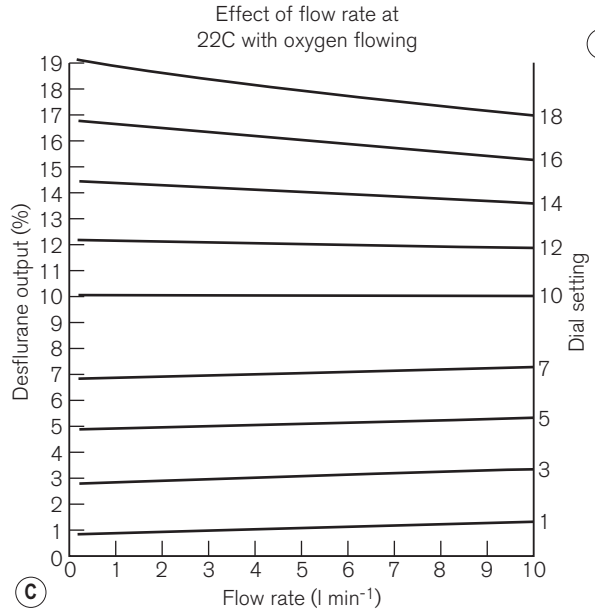


Figure 3.21 **A.** The TEC 6 vaporizer. **B.** Working principles: (1) heater in the vapour chamber, (2) shut-off valve; (3) reservoir, (4) electronic pressure regulator, (5) concentration dial, (6) vaporizer outflow, (7) fresh gas inflow, (8) restrictor, (9) differential pressure sensors. **C.** Performance characteristics of TEC 6 vaporizer.

- There are two more electric heaters in the upper part of the vaporizer to prevent vapour condensation.
- The concentration dial has graduations of 1%, but from 10–18% these are 2% increments. There is an interim stop at 12%, which can be manually overridden to access the higher concentrations.
- The front panel has five LED lights. From top to bottom they are labelled OPERATIONAL to indicate that the unit is ready to be used; NO OUTPUT for when the agent drops below minimum operating level; LOW AGENT to indicate that refilling is required; WARM UP (see below) and

ALARM BATTERY LOW for when the back-up alarm power is either low or disconnected. The latter consists of a 9-volt alkaline battery which requires changing annually. The front panel also houses an LCD (liquid crystal display) of 20 vertically mounted bars that receives electronically processed signals from a sensor in the reservoir. The bars gradually disappear as the vaporizer empties, at which point the heaters are switched off and the low agent LED flashes. There are three symbols displayed on the side of the LCD. The uppermost (equivalent to all 20 bars showing), indicates that the reservoir is full (390 ml). In the middle, a mark indicates that a 240 ml refill is possible (a whole bottle) and the lowest indicates that the reservoir has only 60 ml left.

- At the beginning of the warm-up time, the vaporizer begins a self-testing sequence. The warning alarm sounds for 1 s and all the LEDs flash. When operating temperature is reached, the warm-up light (amber) extinguishes, the operational light glows (green) and the concentration dial unlocks.
- There is a detector which shuts off the vaporizer if it senses more than a 15° tilt off the vertical axis.
- The filler port accepts only the specific filler nozzle (SAF-T-FIl), which is crimped onto the supply bottle for desflurane. To fill the vaporizer, the filler nozzle is pushed into a spring-loaded aperture in the filler port, which is then rotated upwards by inverting the bottle. The contents of the bottle will then decant into the vaporizer reservoir. If the latter is filled only when the LCD bars fall below the 240 ml refill mark, then it will accept the whole bottle. When empty, the bottle may be returned to the starting position at which point the spring in the filler port will eject the filler nozzle. The filling process may be carried out even when the vaporizer is in use. As the bottle is pressurized, it is coated in plastic to prevent the glass splintering in the event of damage. Overfilling is prevented in normal circumstances by placing the outlet from the reservoir above the level attained by the bottle in its filling position. However, should the vaporizer be tilted (and this can only happen if the vaporizer is not in use and not attached to the back bar), overfilling can occur, although liquid will be prevented from leaving the reservoir by the shut-off valve. When the vaporizer is next commissioned, a small amount of liquid might leave the reservoir but would rapidly vaporize.

Dräger D Vapor

This vaporizer (Fig. 3.22) is another example of a measured flow vaporizer for desflurane and works on similar principles to the Tec 6 above. The agent is heated in a chamber so that it behaves as a gas. Its pressure is then regulated electronically to match the patient gas pressure as this alters with changes in flow rate. It is then fed into

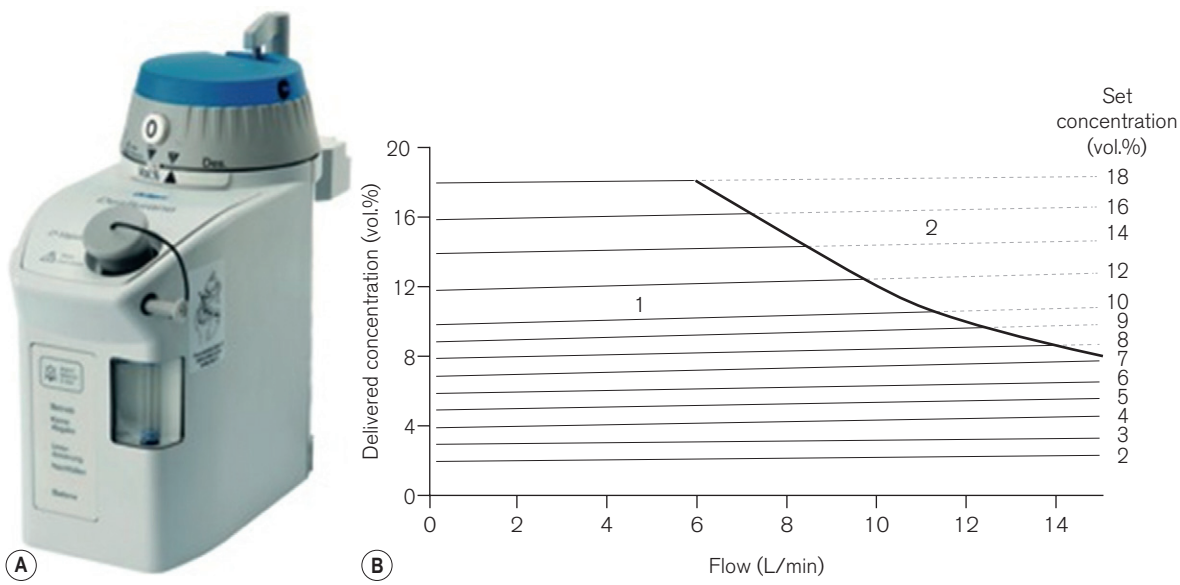


Figure 3.22 Dräger D Vapor **A.** Vaporizer. **B.** Performance characteristics (In area 2 with high gas flows and high dial settings delivered concentrations of desflurane are significantly below those indicated on the control dial.).

the gas stream via the calibrated control knob according to the desired concentration. It is fitted with the ubiquitous Saf-T-Fil connector for desflurane.

The Dräger DIVA

The DIVA (direct injection of volatile anaesthetic) anaesthetic vaporizer is yet another example of a measured flow vaporizer (Fig. 3.23). But unlike the examples above, it has two sections, a plug in vaporizing (and metering) module specific for a particular agent, and a built-in gas supply module that is built into the Dräger Zeus anaesthetic workstation and can accommodate two vaporizing modules.

The vaporizing module has an agent-specific filling device connected to a storage tank, which contains a capacitive level gauge and an overpressure relief safety valve that opens at 1.5 bar. The latter activates only if the metering module is overheated, e.g. by being exposed to direct sunlight, and causing the vapour pressure to rise.

In operation, the pressure of the storage tank can be relieved by electrically operated solenoids within the unit during filling so that it can be safely filled at any time without risk of anaesthetic vapour escaping. The following is a brief and truncated description of the mode of operation. The anaesthetic in the storage tank flows by gravity into the pump tank. When this is full, it is pressurized by air at 2.4 bar from the workstation, forcing liquid agent into a third (metering) tank. From here it passes through an injection valve (actually a fuel injector as used in a Volkswagon Polo) into a heated evaporation chamber that produces saturated vapour. The tanks are connected by one-way valves to prevent back flow. When the pump tank needs refilling, the pressure is reduced to atmospheric to allow refilling from the storage one and then re-pressurized to provide a continuous supply of liquid agent to the vaporizing chamber. The vapour passes through a heated flow sensor in the metering unit into the patient gas circuit. Microprocessor-controlled valves linked to the flow sensor regulate the volume of agent delivered. There are numerous safety features built into the system that are not mentioned here.

Figure 3.23 Dräger DIVA vaporizing modules.

FURTHER READING

Pumping effect

Cole JR. The use of ventilators and vaporizer performance. *Br J Anaesth* 1966;**38**:646–51.

Henegan CPH. Vapour output and gas driven ventilators. *Br J Anaesth* 1986;**58**:932.

Loeb R, Santos B. Pumping effect in Ohmeda Tec 5 vaporizers. *J Clin Monit* 1995;**11**:348.

Thymol and halothane

Carter KB, Gray WM, Railton R, Richardson W. Long term performance of TEC vaporizers. *Anaesthesia* 1988;**43**:1042–6.

Gray WM. Dependence of the output of a halothane vaporizer on thymol concentration. *Anaesthesia* 1988;**43**:1047–9.

Vapour output

James MF, White JF. Anesthetic considerations at moderate altitude. *Anesth Analg* 1984;**63**:1097–105.

Jones CS. Gas viscosity effects in anesthesia. *Anesth Analg* 1980; **59**:92–6.

Leigh JM. Variations on a theme splitting ratio. *Anaesthesia* 1985;**40**:7072.

Loeb RG. The output of four modern vaporizers in the presence of helium. *Can J Anaesth* 1992;**39**:888–91.

Palayiwa E, Hahn CEW, Sugg BR. Nitrous oxide solubility in halothane and its effect on the output of vaporizers. *Anaesthesia* 1985;**40**: 415–19.

Palayiwa E, Sanderson MH, Hahn CE. Effects of carrier gas composition on the output of six anaesthetic

vaporizers. *Br J Anaesth* 1983;**55**: 1025–38.

Schaefer HG, Farman IV. Anaesthetic vapour concentrations in the EMO system. *Anaesthesia* 1984;**39**:171–80.

Scott DM. Performance of BOC Ohmeda Tec 3 and Tec 4 vaporisers following tipping. *Anaesth Intensive Care* 1991;**19**:441–3.

White DC. Symposium on anaesthetic equipment. Vaporization and vaporizers. *Br J Anaesth* (Review) 1985;**57**:658–71.

Wright D, Brosnan S, Royston B, White D. Controlled ventilation using isoflurane with an in-circle vaporiser. *Anaesthesia* 1998;**53**:650–3.

Filling devices

Richardson W, Carter KB. Evaluation of keyed fillers on TEC vaporizers. *Br J Anaesth* 1986;**58**:353–6.

Sato T, Oda M, Kurashiki T. A new agent-specific filling device for anesthetic vaporizers. *Anesthesiology* 1988;**68**:957–9.

Uncles DR, Conway NE. Key issues in vaporizer filling. *Can J Anaesth* 1994; **41**:878–9.

Wittmann PH, Wittmann FW, Connor J, Connor T. A new keyed vaporizer filler. *Anaesthesia* 1994;**49**: 710–12.

Individual vaporizers

Blease Datum User and Maintenance Manual. Issue 2, December 1999.

Dräger Vapor 2000 Instructions for Use, 10th edn. August 2003.

Graham S. The desflurane Tec 6 vaporizer. *Br J Anaesth* 1994;**72**: 470–3.

Hendrick JF, De Cooman S, Deloof T, Vandeput D, Coddens J, De Wolf AM. The ADU vaporizing unit: a new vaporizer. *Anesth Analg* 2001;**93**:391–5.

The TEC 6 Operation and Maintenance Manual. *Datex Ohmeda English Language* Issue 7, March 1994.

The TEC 5 Continuous Flow Vaporizer Operation and Maintenance Manual, *Datex Ohmeda* Issue 1, November 1989.

Penlon Sigma Elite User Instruction Manual Penlon Ltd. February 2001.

Vaporizers in circle systems

Baker AB. Low flow and closed circuits. *Anaesth Intensive Care* 1994;**22**: 341–2.

Brosnan S, Royston B, White D. Isoflurane concentrations using uncompensated vaporisers within circle systems. *Anaesthesia* 1998;**53**: 560–4.

Hazards

Cartwright DP, Freeman MF. Vaporisers (editorial). *Anaesthesia* 1999;**54**: 519–20.

Kharasch ED, Subbarao GN, Cromack KR, Stephens DA, Saltarelli MD. Sevoflurane formulation water content influences degradation by Lewis acids in vaporizers. *Anesth Analg* 2009;**108**:1796–802.

MRHA Medical Device Alert. *Overfilling of Vaporisers* 31, Nov 2003.

Munson WM. Cardiac arrest: hazard of tipping a vaporiser. *Anesthesiology* 1965;**26**:235.

Palayiwa E, Hahn CEW. Overfill testing of anaesthetic vaporizers. *Br J Anaesth* 1995;**74**:100–3.

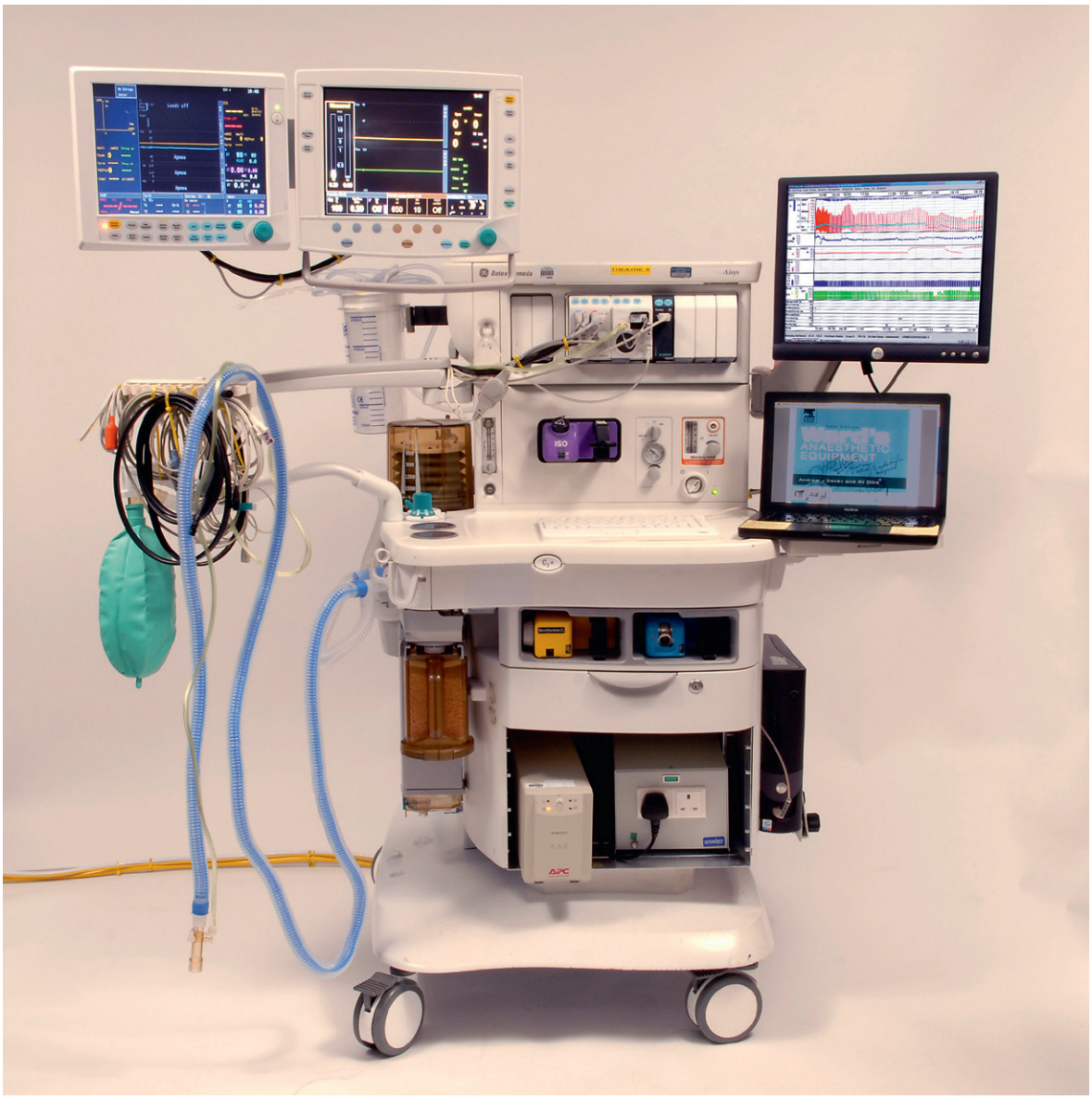


Figure 4.1 Aisys anaesthetic workstation (GE Healthcare, USA), with patient monitoring and additional computer for networked automated record keeping.

modifications having been incorporated, such that the 'anaesthetic machine' component of a modern workstation is still usually recognizable as a descendant of the first-generation Boyle's trolley. Over time, technological safety features have become enshrined in national and international standards which are largely adhered to. As such, an analysis of critical incidents may help to inform a logical approach to the understanding of the safety features and design of modern machines, and for this reason such a section is included in this chapter.

One notable change over the last few years has been the narrowing gap between anaesthesia ventilators and intensive care ventilators. Software-driven, electronically controlled devices are seen that, if not identical in technology to the ITU types, are at least capable of emulating most ventilatory modes if desired. Simple purely mechanical ventilators are seen now only in devices purpose built for the developing countries (see Chapter 27). The reader is referred to Chapters 9 and 10 for more detailed information on ventilators.

FUNCTIONS OF THE MODERN WORKSTATION

Inhalational anaesthesia is still the most commonly used technique worldwide. A few years ago, the AneoTivas, a much simpler anaesthetic machine designed for the delivery of two intravenous anaesthetic agents only (with integral respiratory support and patient monitoring) failed to get past the prototype stage. The anaesthetic workstation, itself an elaboration of the continuous flow anaesthetic machine, has thus developed to accurately and continuously deliver a safe mixture of gasses and vapours for the administration of anaesthesia. The component parts of the modern workstation represent its various and extended functions:

- Safe provision, selection and delivery of anaesthetic gasses and vapours together with an appropriate built-in breathing system, usually a circle system
- Provision of back-up supplies of gasses in the event of failure of the primary sources
- Respiratory support in the form of sophisticated automatic ventilators capable of managing the full range of patient needs
- Monitoring of machine function and settings
- Monitoring of patient physiological variables
- Automated archived record keeping (or at least the facility to output monitored parameters to other systems) and ideally on-line access to information and administration networks
- The integration of the display and auditory signalling of monitored modalities
- Provision of appropriate connection to an anaesthetic gas scavenging system
- Provision of medical vacuum via a suction regulator
- Provision of supplemental oxygen using an auxiliary O_2 flowmeter (for reduced risk of inadvertent anaesthesia)
- Work surface and storage facility for 'everyday' items
- Provision of mains electricity sockets for low-consumption items used in association with delivery or monitoring of anaesthesia, e.g. syringe pumps. (international standards preclude the powering of one medical device from another. For the purpose of this standard, the workstation is seen as a resuscitation trolley.)

DEVELOPMENT OF THE ANAESTHETIC WORKSTATION

The overriding principle governing design of the anaesthetic workstation has been to increase safety in anaesthesia. The invisible and odorless nature of the main gasses used has meant that the focus of safety in the machine has had to be prevention of the accidental delivery of incorrect gasses and gas mixtures. This commences with gas-specific connections to wall and cylinder supplies and continues through non-interchangeable gas-specific pipework within the machine and from there on to standardized arrangements of flow control valves. Along the way, fail-safe devices prevent delivery of nitrous oxide (N_2O) in the event of failure of the oxygen (O_2) supply which is given the highest alarm priority. Carbon dioxide (CO_2), another ill wind, has been largely eradicated from anaesthetic machines, further reducing potential error sources.

Integrated and modular designs

The first machines were solely for gas and volatile agent delivery. Monitoring was a purely clinical modality and a function of the anaesthetist. As individual monitors became available they were connected to or placed onto the machine with a view to creating effectively what would now be termed an *anaesthesia workstation*. The next generation of modern devices briefly attempted to integrate these parts into one harmonious unit (such as the Narkomed 2 and 3 series from North American Dräger discussed in the 4th edition of this text, and below).

Since then though, the design of the anaesthesia workstation has moved away from an integrated single entity towards a modular approach of combining various component parts, perhaps even from different manufacturers, to produce devices adapted for many applications and with less in-built obsolescence. This has been necessitated by the unforeseen growth in the range of possible monitoring modalities which no manufacturer could hope to encompass in one device, and the expansion of the functions of the anaesthesia workstation.

Twenty years ago, when the majority of machines consisted of a stainless steel trolley with a number of monitors stacked on top, at least one manufacturer was making an integrated anaesthesia workstation where gas delivery and patient monitoring were one singular entity. The Narkomed 2a had two CRT screens showing all settings and monitored variables. However, by today's standards, the in-built monitoring was rudimentary. Indeed, it is difficult to imagine how we could once again have a scenario where one in-built monitor can satisfy all requirements unless it has the facility for individual monitoring modalities to be interchanged.

The answer has been to return to a modular approach. The basic pneumatic anaesthetic machine is contained within a chassis or framework to which is added:

- a circle breathing system and usually a 'bag in bottle' arrangement to separate respiratory gasses from the usually pneumatically driven ventilator (see Dräger Zeus for an alternative configuration)
- a choice of ventilator, often a single design but software driven and available in a variety of software configurations

- secondary regulators for damping surges in machine working pressure (see below)
- pressure gauges to show pipeline and cylinder pressures
- a method of metering (e.g. flowmeters), using adjustable valves for proportioning and mixing the various gasses
- a system for attaching vaporizers to the anaesthetic machine for the addition of volatile anaesthetic agents to the gas mixture
- a safety mechanism to warn of the failure of the oxygen supply and to prevent hypoxic mixtures of gas/vapour reaching the patient (oxygen failure warning device)
- a safety mechanism for releasing high-pressure build-up of gasses (back bar pressure relief valve) should a fault occur in the machine
- a system that bypasses the flowmeter for the administration of a high flow of pure oxygen in an emergency
- in-built connection to a circle breathing system with the facility to switch to a single outlet for delivering the gasses and vapours to any other breathing system (the common gas outlet).

Machine framework

The machine framework consists of box-shaped sections of either welded steel or aluminium, which provides both strength and ease of assembly. The design allows for upgrading from a simple model to one with integral monitoring and a ventilator. The machine is usually mounted on wheels with antistatic tyres. These conduct away any static electricity which may affect flowmeter performance and which also present a risk of ignition of flammable anaesthetic agents (in parts of the world where these may still be used).

The compressed gas attachments

Compressed gasses to be used by the anaesthetic machine (air, oxygen and nitrous oxide) are usually supplied by two methods: cylinders and pipelines.

Cylinders

The cylinders are clamped onto the machine by a yoke arrangement and secured tightly using a wing-nut (Fig. 4.2). To prevent installation of the wrong gas cylinder to a yoke, the cylinder heads are coded with appropriately positioned holes that match pins on the machine yoke. This is called a pin index system, for which there is an internationally agreed standard (ISO 407:2004) (see Fig. 1.5). A thin neoprene and aluminium washer (Bodok seal) is interposed between the cylinder head and yoke to provide a gas-tight seal when the two are clamped together.

Cylinder yokes are also fitted with filters and one-way spring-loaded non-return (check) valves. The one-way valve prevents retrograde leaks when a cylinder is removed.

A leak of not more than 15 ml min^{-1} through an open yoke is acceptable in new machines. However, in older machines the non-return valve may not be as efficient owing either to the design (the valve not being spring loaded) or to wear and tear, and could result in greater than acceptable backpressure leaks. These leaks, when occurring unexpectedly, have been shown to alter the composition of the gas leaving the flowmeter block and have resulted in the delivery of a hypoxic gas mixture to an attached breathing system (see section on Flowmeters). Blanking plugs (dummy cylinder heads) are available and

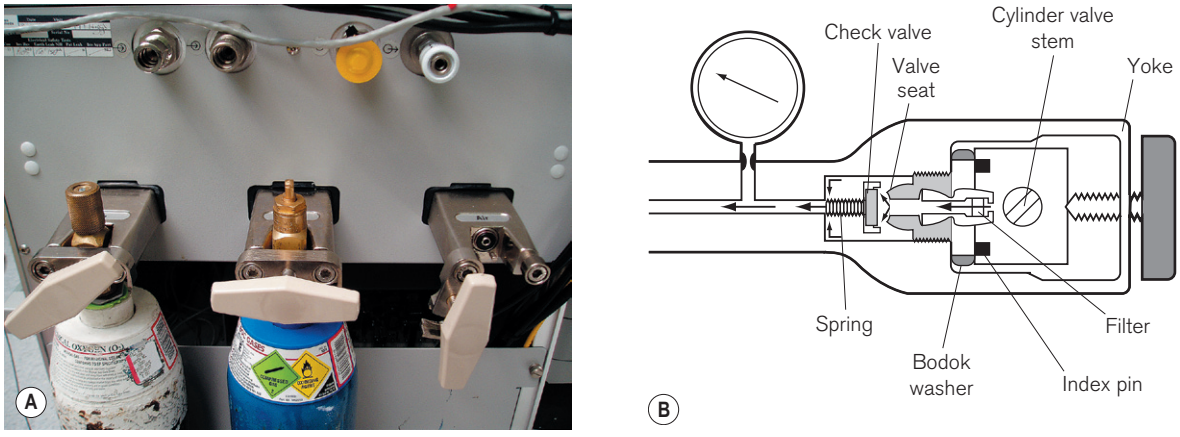


Figure 4.2 **A.** Cylinder yokes. The empty right-hand yoke shows a Bodok seal and the pins of the pin index system. **B.** Diagram of cylinder yoke assembly.

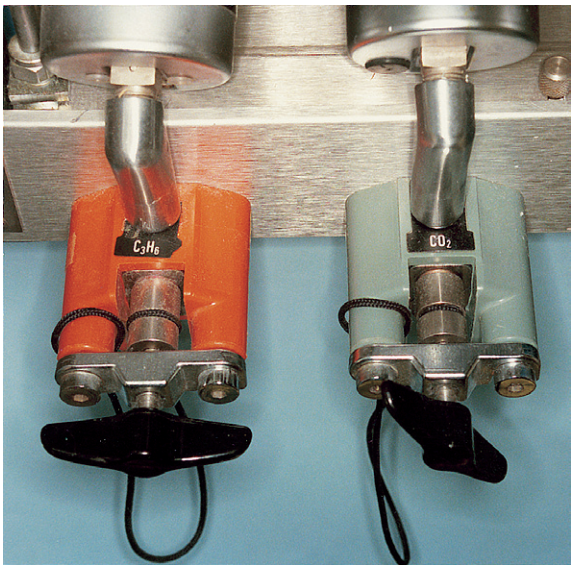


Figure 4.3 Blanking plugs on somewhat ancient cyclopropane and CO_2 cylinder yokes.

should be inserted into all empty yokes to overcome this problem (Fig. 4.3).

Pipelines

Each pipeline source is attached to the machine via a gas-specific connection. In the UK this is called a NIST (non-interchangeable screw-threaded) connection and is discussed in greater detail in Chapter 1 (see Fig. 1.29). In the USA, a similar system is employed called DISS (Diameter Indexed Safety System). However, the diameters of the nipples and bodies for the various connections are smaller and not compatible with the NIST system.

The *pipeline union block* (the point of attachment of the pipeline to the machine) usually contains a metal gauze filter and also a one-way spring-loaded check valve to prevent retrograde gas leaks should the relevant system be disconnected.

Pressure (contents) gauges

The pressure in cylinders and pipelines is measured by Bourdon-type gauges (see Fig. 2.6). The gas entry to the pressure gauge has a constriction so as to smooth out surges in pressure that could damage the gauge, as well as to prevent total and rapid loss of gas should a gauge rupture. The gauges are labelled and colour coded for each gas, according to the standards for each country. They are also calibrated for each gas used on the machine. The scale on the gauge extends to a pressure at least 33% greater than either the filling pressure of the cylinder or pipeline pressure (at a temperature of 20°C). Each cylinder yoke

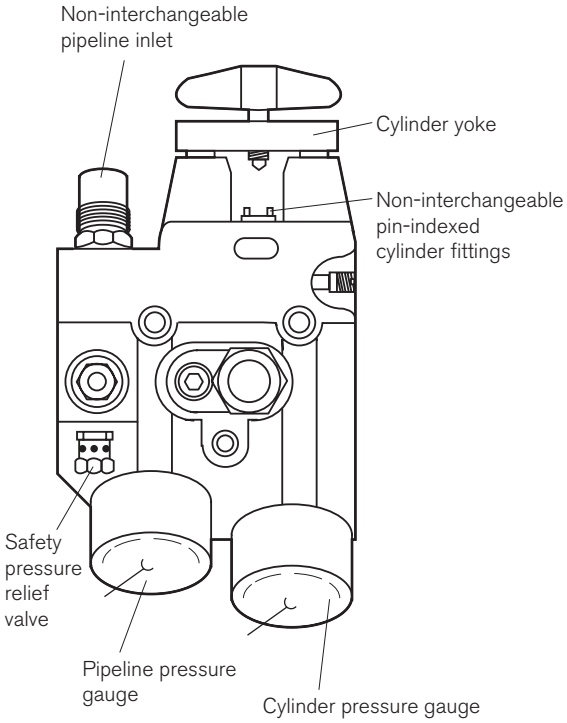


Figure 4.4 Single-cast gas block.

may be fitted with a gauge or, alternatively, a single-cast brass block may be used to house the NIST/DISS pipeline connection, cylinder yoke, pressure regulator and housings for the pressure gauges (Fig 4.4). This minimizes the number of connections and potential leaks, but is a somewhat dated arrangement now.

Single-block manifold

The single-cast brass block has given way in recent years to the single-block gas manifold (Fig. 4.5), which is drilled to accommodate as many as possible of the gas pipeline and pneumatic valve interconnections. The small, compact manifold (Fig. 4.5A), for reasons of design economy, does not carry the cylinder yokes and is hence typically downstream of the primary pressure regulator which is invariably part of the cylinder yoke assembly (Fig. 4.5B).

Pressure regulators (reducing valves)

Pressure regulators are used on anaesthetic machines for three main reasons:

1. The pressure delivered from a cylinder is far too high to be used with safety in apparatus where a sudden surge of pressure might accidentally be delivered to the patient.

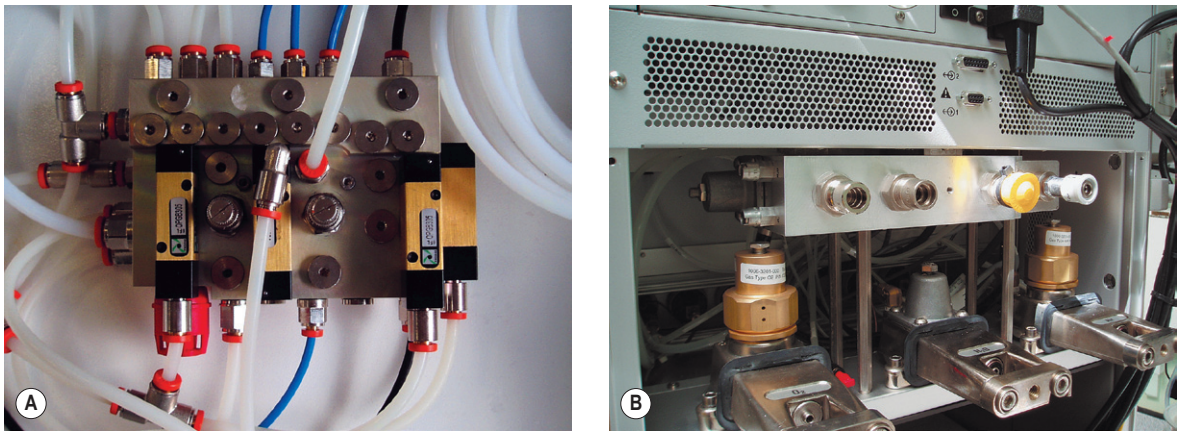


Figure 4.5 **A.** Single-block manifold arrangement of Blease anaesthesia machine. The coil of nylon tubing just visible is an ingenious approach to the need for an O_2 reservoir for operating the Ritchie whistle. **B.** Datex-Ohmeda Aestiva with back plate removed demonstrating cylinder yokes and primary pressure regulators for O_2, N_2O and air. Above these is the aluminium single-block manifold carrying NIST connectors for O_2 and N_2O pipeline supplies (air inlet is blanked off) and auxilliary high-pressure oxygen outlet.

2. If the pressure were not reduced, flow-control (fine-adjustment) valves, tubing and various other parts of the apparatus would have to be much more robust, and a fine and accurate control of gas flow would be difficult to achieve. There would also be a danger of pressure building up and damaging other components of the apparatus.
3. As the contents of a cylinder are exhausted, the pressure within the cylinder falls. If there were no regulating mechanism to maintain a constant reduced pressure, continual adjustment would have to be made of the flow-control valve in order to maintain a given flow rate.

Not only is the pressure reduced, but it is also kept constant, and for this reason this type of valve is more correctly termed a *pressure regulator.*

Working principles

In Fig. 4.6, the chamber C is enclosed on one side by the diaphragm D. As gas enters the chamber through the valve V, the pressure in the chamber is increased and the diaphragm is distended against its own elastic recoil and the force from the spring S. Eventually the pressure rises high enough to move the diaphragm far enough to close valve V. The pressure at which this occurs may be varied by adjusting the screw X so as to alter the force exerted by the spring S. If gas is allowed to escape from the outlet of the chamber, the pressure falls and valve V reopens. When the regulator is in use, a steady pressure is maintained in the chamber by the partial opening of valve V.

In another form of regulator (Adams valve), the push-rod is replaced by a 'lazy tongs' toggle arrangement

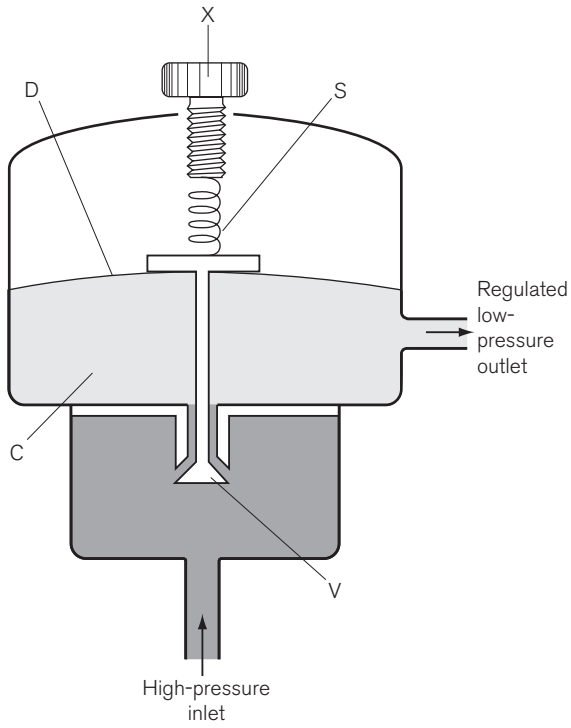


Figure 4.6 A simple pressure regulator. D, diaphragm; S, spring; C, low-pressure chamber; V, valve seating; X, adjustment screw.

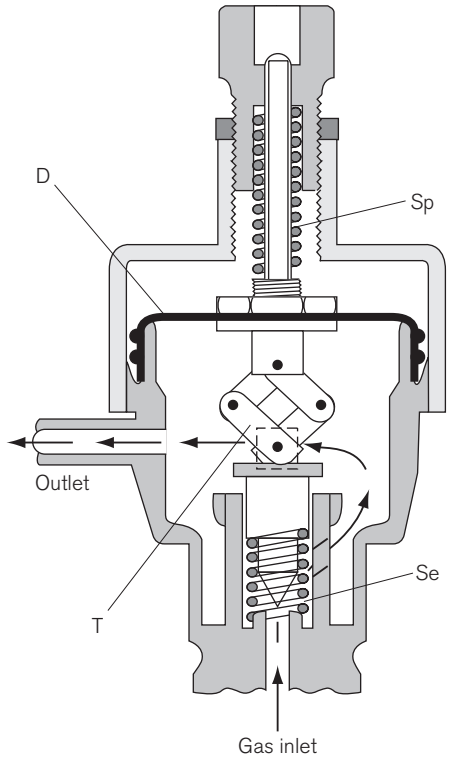


Figure 4.7 The Adams regulator. D, diaphragm; Sp, spring; Se, seat; T, toggle levers.

(Fig. 4.7), which reverses the direction of the thrust transmitted from the diaphragm.

The accuracy of regulators

Let us consider that the push-rod is pushed downward by two forces: the compression in the spring and the elastic recoil of the diaphragm (Fig. 4.8). Let these be added together and represented by *S*. The force that opposes *S* consists of two parts: the high pressure (*P*) of the gas pushing on the valve V over an area of *a*; and the low pressure (*p*) acting on the diaphragm over an area *A*, so:

$$S = Pa + pA$$

Thus, if *S* remains constant, as *P* falls, *p* rises so that as the cylinder empties, the regulated pressure increases. In fact, as *P* falls, the valve V will have to open further to permit the same flow rate. The spring expands and therefore its compression is reduced, and in the same way the tension in the diaphragm is reduced. Therefore, as *P* falls, there is a small reduction in *S*, which partially reverses the effect illustrated here.

In the Adams valve (Fig. 4.9), it can be seen that the pressure *P* exerted by the high-pressure gas to open valve V is assisted by the spring and the recoil of the diaphragm *S*. These forces jointly oppose the force exerted by the low-pressure gas on the diaphragm, so:

$$Pa + S = pA$$

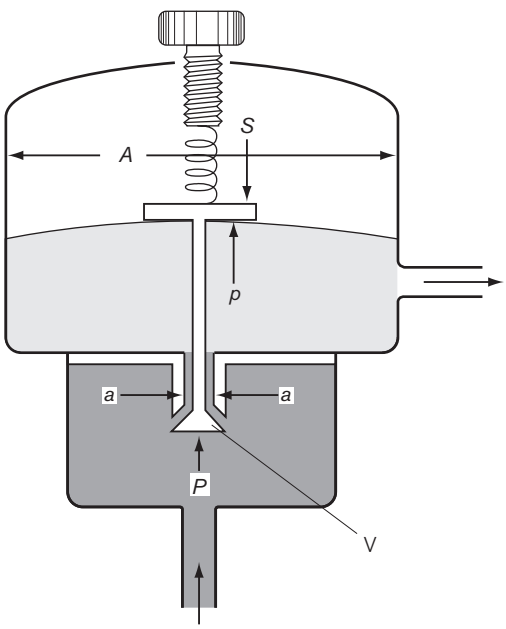


Figure 4.8 Forces acting in a simple regulator.

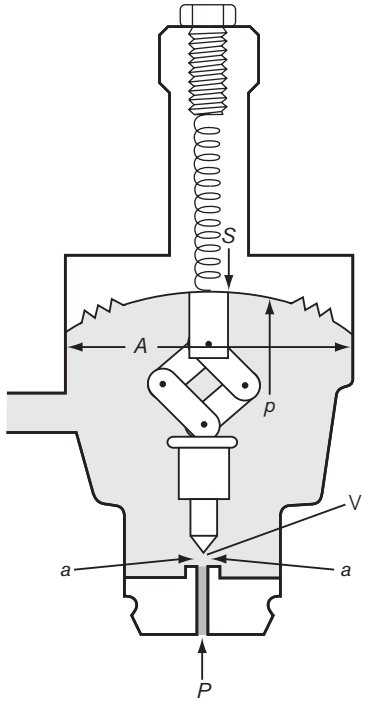


Figure 4.9 Forces acting in an Adams regulator.

Now as P falls, so does p; therefore the regulated pressure falls slightly as the cylinder pressure drops. At the same time the valve V opens slightly and this, by allowing the spring to expand, reduces S, which slightly accentuates the fall in p. The fall of p can be minimized by making S great compared with Pa.

There are several types of pressure regulator available, the choice being dependent on:

- the maximum flow rate required
- the regulated pressure to which it is to be set
- the maximum input pressure that it is to handle.

For low-pressure regulators, the diaphragms are frequently made of rubber or neoprene, whereas in those for higher pressures the diaphragm is made of metal. Adjustments to alter the regulated pressure should be made only by service engineers. On some anaesthetic machines, 'universal' regulators are used. These operate equally well from an input of 420 kPa (60 psi) from the pipeline, as from a maximum of 20 000 kPa (2900 psi) from cylinders and are of the Adams type. The term 'universal' is also used in a different context, as below.

Interchangeability of regulators

Pressure regulators used to be labelled and coded for specific gasses. This is because a special alloy was required in the valve seating for some gasses (e.g. nitrous oxide) in order to prevent corrosion. However, modern regulators are designed to be compatible with all anaesthetic gasses. This is achieved by using materials such as PTFE coatings on the diaphragms together with nitrile (a hard synthetic rubber) valve seats and chrome-plated brass for the regulator body (Fig. 4.10). These too are described as 'universal' by their manufacturers.

Common faults in regulators

- Damage to the soft seating of valves may occur as a result of the presence of grit or dust, usually from a dirty cylinder. This may cause a steady build-up of pressure in the apparatus when the cylinder is left turned on but with no gas flowing.
- A hissing noise may indicate a leaking or burst diaphragm. The regulator will need replacing or repairing by the manufacturer or service engineer.
- Adams valves sometimes develop a fault that causes continual 'jumping' of the flowmeter bobbin, indicating an intermittent change of pressure and flow rate. This is usually due to the 'lazy tongs' sticking as a result of wear, but it may also be caused by small particles of grit or metal in the lazy tongs or the valve seating.

The older versions of the Adams valves had fins on their nitrous oxide regulators to conduct heat from the surrounding air to prevent excessive cooling of the valves. Prior to this, it was not uncommon for the nitrous oxide to contain a significant quantity of water vapour as an impurity, and this condensed onto the valve seating and then froze, thus jamming the valve. The extra heat conducted by the fins was sufficient to prevent this freezing.

Relief valves on regulators

Safety blow-off valves are often fitted on the downstream side of regulators to allow the escape of gas if, by accident, the regulators fail and allow a high-output pressure. With a regulator designed to give a pressure of 420 kPa (60 psi), the relief valve may be set at 525 kPa (70 psi). These valves may be spring loaded (Fig. 4.10), in which case they close when the pressure falls again, or they may operate by rupture, in which case they remain open until repaired. As a further safety feature, a flow restrictor (usually in the form of a simple pin hole orifice) on the high pressure inlet side of the regulator, limits maximal flow from the cylinder to between 70 and 150 l min^{-1}, ensuring that in the event of regulator failure the high-pressure relief valve can dump the maximal flow without further pressure rises.

Primary pressure regulators

Modern anaesthetic machines may have several pressure regulators (primary and secondary) for each gas. Primary regulators are used to reduce high cylinder pressures to lower machine working pressure (typically 420 kPa (60 psi)). The pressures downstream of the primary regulators are not the same in all machines or in all countries.

Some manufacturers adjust their cylinder regulators to just below 420 kPa (60 psi) (i.e. 350 kPa (50 psi)) to give *pipeline preference*. This allows the anaesthetic machine to preferentially use pipeline gas when the reserve cylinders have been accidentally left turned on, so reducing the potential for premature emptying of these cylinders. However, even with this differential, many regulators are known to 'weep', i.e. gradually empty their contents. Reserve cylinders should, therefore, always be turned off after testing until required again.

Secondary pressure regulators

Several factors cause the machine working pressure (420 kPa in the UK) to fluctuate by up to 20%. For example, at times of peak demand in a hospital, pipeline pressures may well drop by this amount. Similarly, if an auxiliary outlet on the anaesthetic machine is used to drive a ventilator with a very high sudden and intermittent gas demand, a similar pressure drop will occur before the pipeline or cylinder is able to restore the supply. These pressure fluctuations produce parallel fluctuations in flowmeter performance. A second (*secondary*) regulator set below the anticipated pressure drop smoothes out the supply, minimizing these fluctuations. This is important in machines incorporating mechanically linked anti-hypoxia systems attached to the flowmeter bank (see below), as these systems assume that the oxygen supply

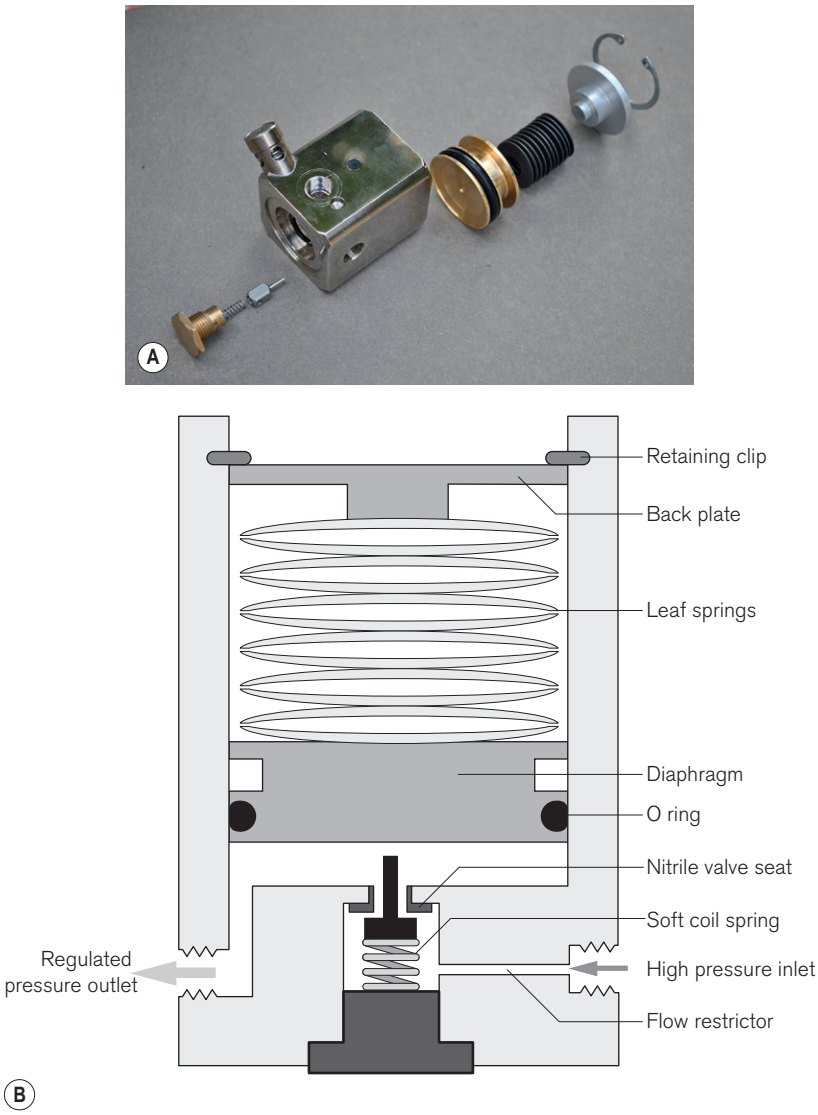


Figure 4.10 Modern Pneupac universal regulator, **A.** disassembled and **B.** diagrammatic representation. The safety valve seen in the photograph (see text, Relief valves on regulators) on the low-pressure chamber is not shown in the diagram.

pressure is constant in order to achieve an accurate flow of gas. A mechanically linked system would not be able to detect altered gas flow rates caused by changing pressures. Furthermore, secondary regulators also prolong the accurate supply of oxygen to the flowmeter if there is a gradual failure of the oxygen supply (i.e. cylinder emptying) prior to the oxygen failure warning device being activated.

Regulators have to meet stringent criteria before being installed. They are required to withstand pressure of 30 mPa (megaPascals) without disruption and their output should not vary more than 10% across a wide flow range (100 ml min^{-1} to 12 l min^{-1}). They should also be fitted with a pressure relief valve that opens at a pressure not exceeding 800 kPa (UK).

Flow restrictors

It was an occasional practice to entirely omit regulators from the pipeline supply (420 kPa (60 psi)) in older anaesthetic machines. Sudden pressure surges at the patient end of the anaesthetic machine were prevented by flow restrictors. These consisted of constrictions in the regulated pressure pipework upstream of the flowmeters. The disadvantage of using flow restrictors without regulators was that changes in pipeline pressure were reflected in

changes of flow rate, which made readjustment of the flow control valves necessary. Also, there was a danger that if there was an obstruction at the outlet from the anaesthetic machine, pressure could build up in the vaporizers and cause damage. This was normally prevented by the inclusion of a 'blow-off' safety valve (see section on 'the back bar' below). Flow restrictors do not normally require any maintenance.

Gas-tight connections within the machine

The various components within the anaesthetic machine are joined to each other by a series of pipes. Although now almost entirely made of high-density nylon, previously copper piping was standard, and is still used occasionally; hence these components are also briefly described below. Piped medical gas conduits within hospital walls and ducting are still entirely of metal.

Whilst there is no standard for the design of gas piping within the machine, with the advent of nylon tubing, manufacturers tend to use pipes of differing diameters and/or sometimes colour for the different gasses to reduce the risk of accidental misconnection during servicing and assembly (see Fig 4.5A where the white tubing is for oxygen and the blue and black, respectively, for nitrous oxide and air).

Joints in metal tubing

These may be permanent or detachable. Two metal pipes may be permanently joined by one of two methods:

1. One pipe may have a slightly larger diameter than the other so that they overlap (Fig. 4.11A).
2. Where the diameters are similar, both ends are inserted into a sleeve of metal (Fig. 4.11B).

The adjacent surfaces are then bonded together by brazing (applying a molten filler alloy whose melting temperature is above 430°C) or hard soldering (a similar principle using an alloy with a lower melting point). After making such a joint it is important that all traces of flux are removed. Flux is a material applied to the surfaces to be bonded, allowing the molten filler to spread more

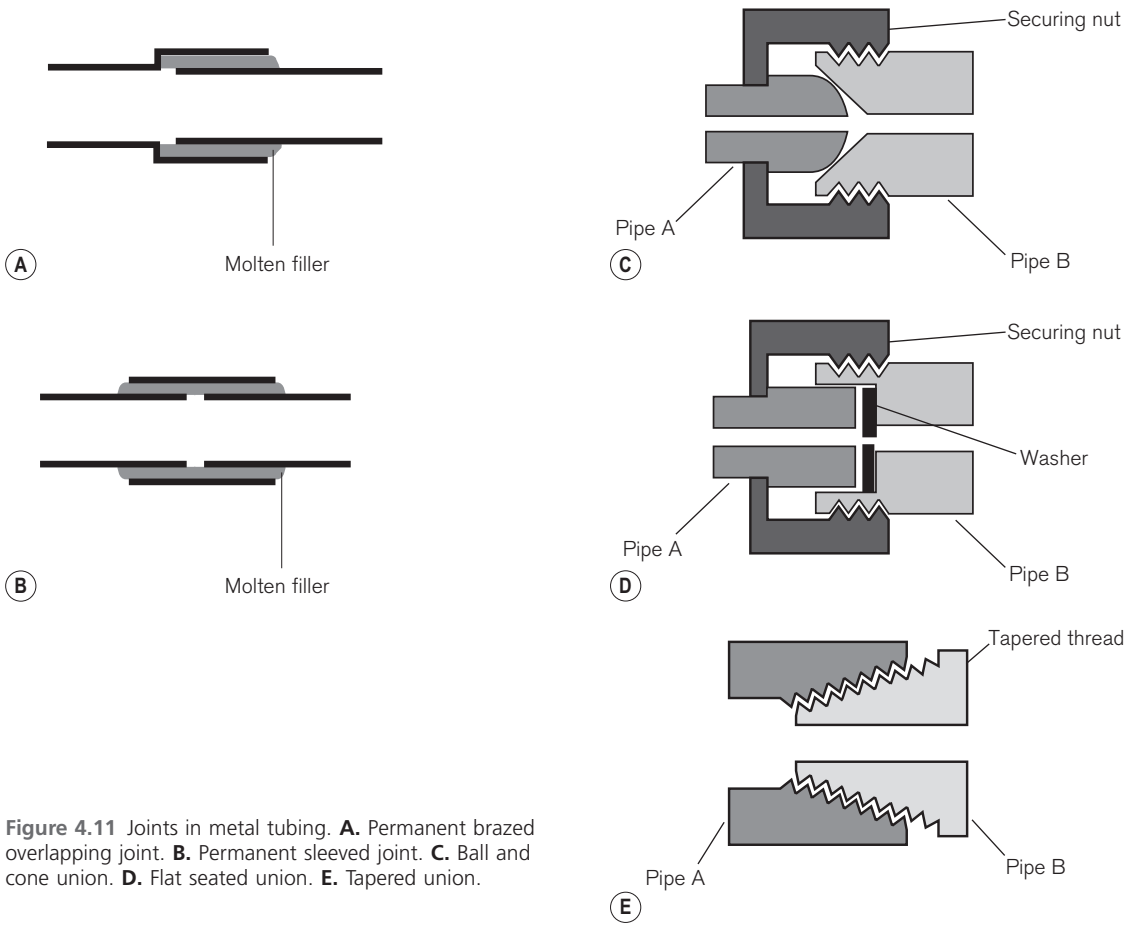


Figure 4.11 Joints in metal tubing. **A.** Permanent brazed overlapping joint. **B.** Permanent sleeved joint. **C.** Ball and cone union. **D.** Flat seated union. **E.** Tapered union.

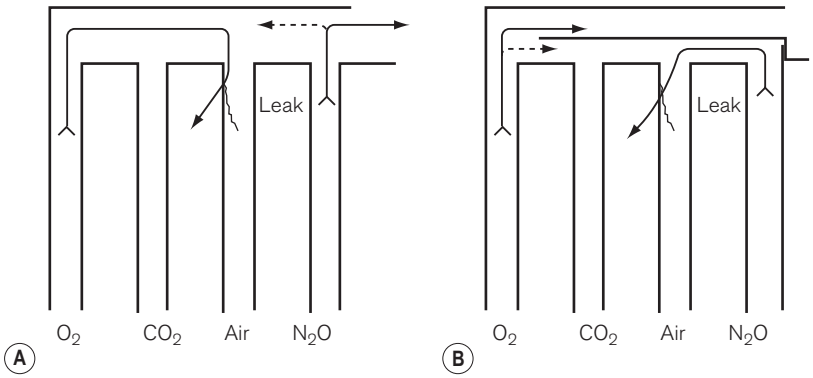


Figure 4.16 Diagram to show the effect of a leak from one of the rotameter tubes. **A.** A leak from the cyclopropane tube in the traditional form of flowmeter block would result in back-pressure from the nitrous oxide, causing oxygen to escape through the leak. The patient would therefore receive a hypoxic mixture. **B.** A rearrangement whereby the oxygen is the last gas to enter the mixed gas flow and nitrous oxide rather than oxygen would be expelled through a leak. This would not lead to the patient receiving a hypoxic mixture.

2. Values must be accurate to within 10% of the indicated flow (at flow rates between 10% of full scale or 300 ml min^{-1}, whichever is the greater, and 100% of the maximum indicated flow).
3. When axial push or pull forces are applied to the valve spindle without rotation (at a flow rate 25% of the maximum indicated flow), the maximum flow change must not be greater than 10% or 10 ml min^{-1}, whichever is the greater. (Much older machines may have spindles that do not meet this requirement. Axial pressures at a flow rate of 1 l min^{-1} have been shown to change the flow rate by 50% in these machines, with resultant hypoxic mixtures being delivered to the patient, when they are used with a low-flow anaesthetic breathing system.)
4. Each flow control valve must be permanently and legibly marked, indicating the gas it controls (using the name or chemical symbol).
5. As well as conforming to (4), the oxygen flow control knob (Fig. 4.14) must have an individual octagonal profile. When the valve is closed the knob must project at least 2 mm beyond the knobs controlling other gasses at all flow rates. Its diameter must also be greater than the maximum diameter of the flow control knobs for other gasses.

The flowmeter block

In the UK and many other countries, the flowmeters are traditionally arranged in a block with the oxygen flowmeter on the extreme left, the nitrous oxide on the extreme right and those for compressed air and carbon dioxide (where fitted) in between these. However, some machines, such as those originally manufactured by Datex-Ohmeda (now GE Healthcare) and Blease (Spacelabs Healthcare), incorporate a system that delivers a minimum concentration of oxygen (e.g. 25%), and requires the oxygen and nitrous oxide flow control valves to be adjacent, as they are linked by a sprocket and chain or cogwheel.

The flowmeters are mounted vertically and usually next to each other, in such a way that their upper (downstream) ends discharge into a manifold. Traditionally, this was unfortunately arranged in such a way that if there were a leak in, say, the central tube, oxygen would be lost rather than nitrous oxide. As a result, a hypoxic mixture could be delivered to the patient (Fig. 4.16A). In most modern machines, oxygen is the last gas to flow into the manifold so that a leak would not lead to such a hypoxic mixture (Fig. 4.16B). As a solution to the same problem, in some countries such as the USA and Canada, the order of the flowmeters in the block has been reversed, with oxygen on the right. However, this too has previously (before the advent of hypoxic guard interlinks) led to patients receiving a hypoxic mixture because anaesthetists have not been made aware of the transposition.

The practice of removing carbon dioxide cylinders (and in the past cyclopropane) from their yokes has exposed a further hazard in older machines. Oxygen can be lost via a retrograde leak through a carbon dioxide (or cyclopropane) flowmeter, even when intact, if the corresponding needle valves are inadvertently left open. Gas can track back from the manifold via the flowmeter and open needle valve to the unblocked cylinder yoke and escape. The one-way (check) valves fitted to cylinder yokes in some machines were never intended to provide a perfect gas-tight seal under all conditions. They were not spring loaded because they were designed to work against high back pressures rather than the relatively low back pressures produced in the retrograde leak. This leak may be increased by adding an extra resistance to flow downstream of the flowmeter block (i.e. some types of minute volume divider ventilator or high-resistance vaporizer), which effectively increases the gas pressure in the

flowmeter block. All empty cylinder yokes for air, carbon dioxide and cyclopropane (where these still exist) should be fitted with blanking plugs (Fig. 4.3) so as to prevent this problem. Modern standards (BS EN 740:1999) stipulate that a maximum retrograde flow of 100 ml min^{-1} is acceptable, or 10 ml h^{-1} if the fault condition causing this is not alarmed.

Recent increased interest in low-flow anaesthesia systems has created a demand for flowmeters that can more accurately measure flows below 1 l min^{-1}. This is achieved by the use of two flowmeter tubes for the same gas. The first is a long thin tube accurate for flows from 0 to 1000 ml min^{-1} that complements the second conventional tube calibrated for higher flows (1–10 l min^{-1} or more). Both are activated from the same flow control valve. These 'cascade' flowmeter tubes for each gas are arranged sequentially (Fig. 4.15) so that when the flow control valve is opened the low-flow tube is seen to register first. At flows over 1 l min^{-1}, the bobbin in the low-flow tube is no longer easily visible.

Carbon dioxide flowmeters

The provision of carbon dioxide on anaesthetic machines is somewhat controversial, as several deaths have occurred owing to the inadvertent and excessive use of the gas. Typically, in these accidents, the flowmeter valve had been left fully open either during a check procedure or at the end of a previous case, and the bobbin was not readily noticed at the top of the flowmeter tube. The next patient then received in excess of 21 min^{-1} of carbon dioxide. Flowmeters have, therefore, been introduced that do not have a bezel which can hide the flowmeter bobbin at the top of the tube. Current standards permit a maximum flow of 600 ml min^{-1} from carbon dioxide flowmeters.

Anti-hypoxia devices

Anaesthesia machines in use now must either not be capable of delivering a gas mixture with less than 20% oxygen or have a means to give an alarm at an oxygen concentration of below 20% in the inspiratory gas which must be separate from any 'add on' patient respiratory gas monitoring (EN 740:1999). Of these approaches it has proven ultimately safer and simpler to design a system whereby it is physically impossible to set the nitrous oxide and oxygen flow rates to give hypoxic mixtures. Some approaches taken by manufacturers are discussed below.

Mechanical devices

The 'Link 25' system (Ohmeda) (Fig. 4.17) incorporates a chain that links the flow control valves for nitrous oxide and oxygen. There is a fixed sprocket (cog) on the nitrous oxide spindle that relays its movement to a larger cog on the oxygen flowmeter spindle via a chain. The oxygen cog moves along a static, hollow worm gear, through which the oxygen flowmeter spindle passes. As the nitrous oxide flowmeter control is turned counter-clockwise (increasing the nitrous oxide flow), the chain link moves this larger cog nearer to the oxygen flowmeter control so that, when a 25% oxygen mixture is reached, it locks on to the oxygen control knob and moves it synchronously with any further increase in nitrous oxide flow. The oxygen flow control can of course be independently opened further, but cannot be closed below a setting that if nitrous oxide is flowing, will produce less than 25% oxygen in the mixture. Other manufacturers use interlinking gears (Fig. 4.18) to achieve the same effect. This type of mechanical link, however, has some limitations:

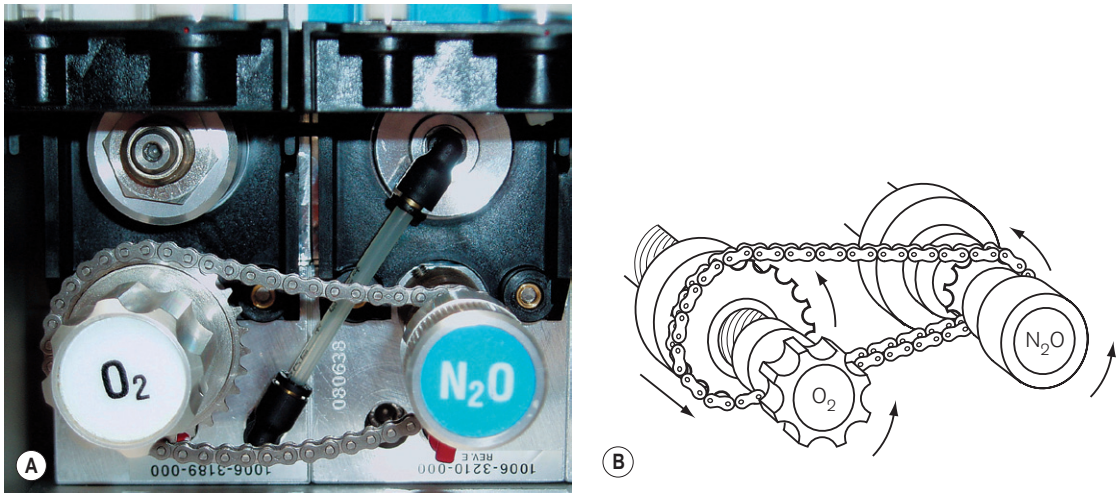


Figure 4.17 'Link 25' system. **A.** In place on the Datex-Ohmeda Aestiva/5. **B.** Schematic of operation.



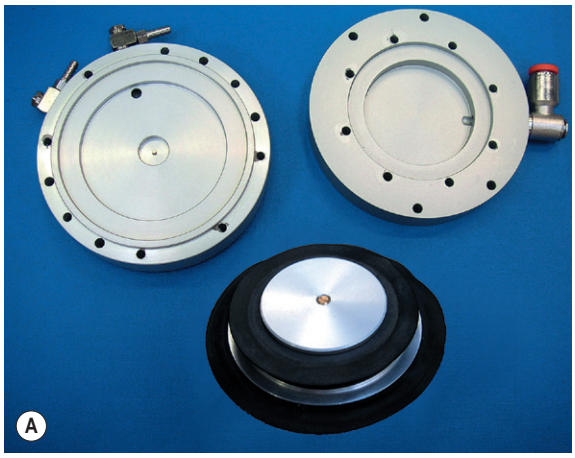
Figure 4.18 Blease mechanical interlink hypoxic guard, showing stop for minimum O_2 flow.

- It takes no account of other gasses in the flowmeter block (air and carbon dioxide) that could potentially dilute the mixture below a 25% oxygen concentration.
- On its own it will not recognize and compensate for variations in gas supply pressure that affect flowmeter performance. However, these systems include secondary pressure regulators (see above) in both the oxygen and nitrous oxide systems, the purpose of which is to prevent variations in gas supply pressure from affecting flowmeter performance.
- At low fresh gas flows into a circle system, a 25% oxygen ratio may be insufficient to prevent the circle patient gas mixture from becoming hypoxic (see Fig. 5.22). Hence a minimum basal flow of oxygen (see below) or a 50% oxygen ratio at low flows is required.

A further safety feature of this system includes a mechanical stop fitted (Fig. 4.18) to the oxygen flowmeter control valve, ensuring that a preset minimum standing flow (typically between 25 and 250 ml min^{-1}) of oxygen is maintained even when the valve is fully closed. This flow, of course, can occur only when the machine master switch for all the gasses is switched on.

Pneumatic devices, e.g. Pneupac ratio system

Originally designed and launched by Medical and Industrial Equipment Ltd UK (MIE) in 1988, this system relies on a ratio mixer valve (Fig. 4.19) to ensure that the oxygen concentration leaving the flowmeter block never drops below 25% of the nitrous oxide concentration. When the machine master switch is turned on, a basal flow rate of 200–300 ml min^{-1} of oxygen is established (Fig. 4.20A). This is independent of, and bypasses, the ratio mixer valve. Nitrous oxide supplied to one side of the valve exerts a pressure on the diaphragm which is opposed by the pressure exerted by oxygen on a separate but coupled larger



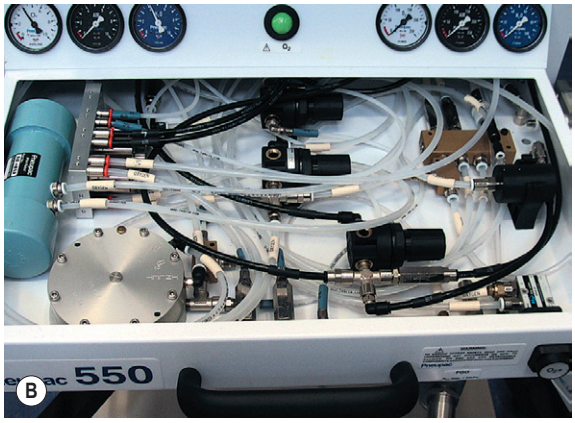


Figure 4.19 Pneupac oxygen ratio system valve **A.** opened (and image manipulated) to better demonstrate double diaphragm arrangement and their relative sizes, note that a spacer ring that is interposed between the two valve housings is not shown here **B.** in place in a Pneupac anaesthesia machine.

diaphragm (Fig. 4.20B). Inward movement of the oxygen diaphragm is linked to the opening of a poppet valve that allows more oxygen to flow through the O_2 chamber thus balancing the opposing forces. Any increase in the flow of nitrous oxide results in an increase in pressure on the nitrous oxide side of the diaphragm, causing the latter to move towards the compartment containing oxygen. The ratio of the area of the diaphragms is so constructed that the oxygen flow rate will increase by a ratio of 25% of any increase in the nitrous oxide flow rate. This increased oxygen flow is independent of the main oxygen flow control valve that bypasses the ratio mixer valve and, of course, can be adjusted independently. The ratio mixer valve does not work in reverse as it takes a single passive N_2O connection from downstream of the N_2O flow control valve whilst taking O_2 from upstream (before) of the O_2

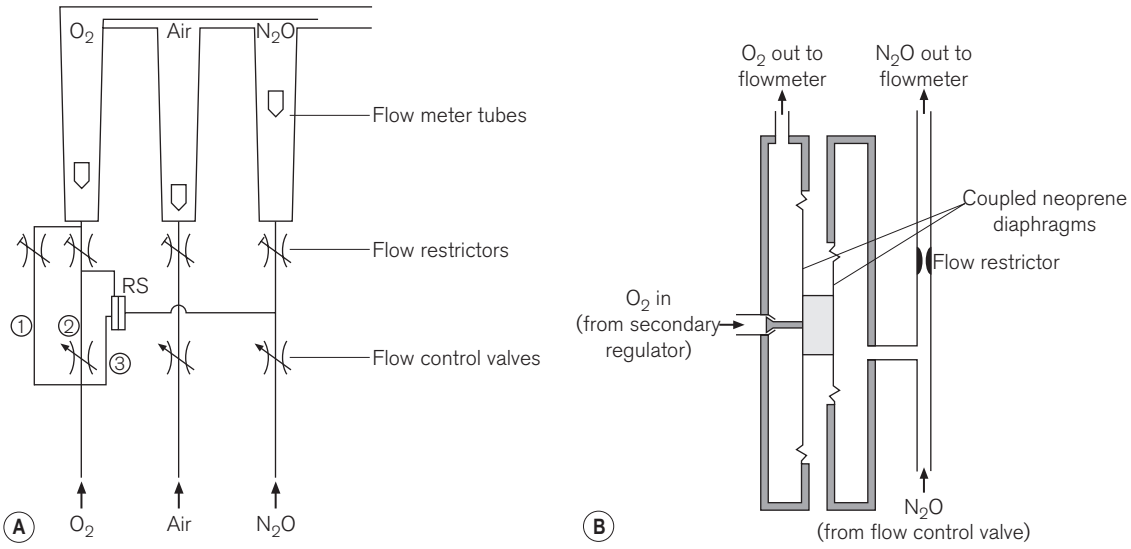


Figure 4.20 Pneupac oxygen ratio system. **A.** Schematic of pneumatic connections. Abbreviations: (1) independent basal flow set at 250 ml min^{-1}; (2) operator variable flow from flow control valve; (3) additional flow through oxygen ratio system (RS) to make total O_2 flow equivalent to at least 25% of N_2O flow. **B.** Schematic of oxygen ratio system showing coupled diaphragms.

flow control valve; hence, if the nitrous oxide flow rates are reduced, the oxygen flows remain as set before by the O_2 flow control valve. The double diaphragm arrangement also means that rupture of a diaphragm will not result in contamination of the O_2 flow by N_2O.

Electronically controlled anti-hypoxia devices (Penlon Ltd)

In this system, a paramagnetic oxygen analyzer continuously samples the gasses mixture leaving the flowmeter bank. If the oxygen concentration falls below 25%, a battery-powered electronic device sounds an audible alarm and the nitrous oxide supply is cut off. This results in an increase in the oxygen concentration and, as a result, the nitrous oxide supply is temporarily restored. If the oxygen flow rate has not been increased, the nitrous oxide disabling system is reactivated and the alarm will again sound. The whole process is repeated, thus providing an intermittent oxygen failure alarm and at the same time assuring a breathing mixture with more than 25% oxygen (although the total flow rate will be lower than intended).

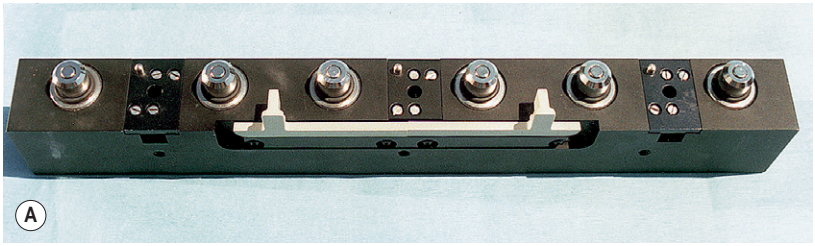
If the oxygen supply fails completely, there is a continuous audible alarm. The power is provided by a maintenance-free lead-acid battery that is kept charged by the mains electricity supply while the machine is in use and will continue to operate in the absence of a mains supply for 1.5 h. If the audible alarm is activated during this period it will sound for 20 min, after which a visual and audible 'low battery' warning is given. If for some reason the lead-acid battery is not adequately charged at the beginning of an anaesthetic session, the nitrous oxide supply (as well as medical air in US versions) is disabled and cannot be used. However, under no circumstances is the oxygen supply interrupted. This alarm is in addition to the standard oxygen failure warning device (Ritchie whistle, see below).

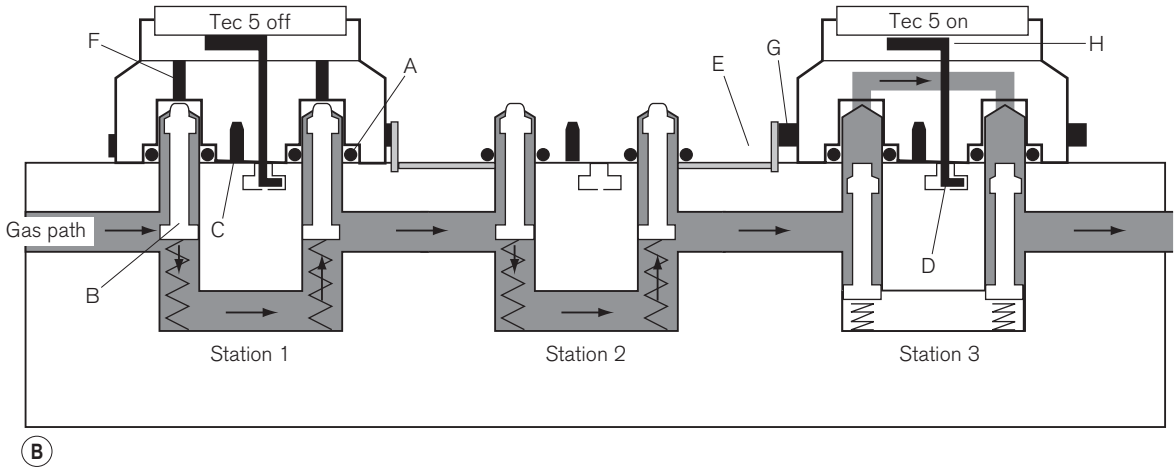
Penlon stopped installing this electronic system of hypoxia protection in 2001, largely for reasons of cost, but many of their machines are still currently in use with this technology. Penlon now use a mechanical interlink. (See below, Specific machines, Dräger Primus for a further example of electronic control of oxygen ratio).

The back bar

Strictly speaking, the term 'back bar' describes the horizontal part of the frame of the machine, which supports the flowmeter block, the vaporizers and some other components. However, it is often used loosely to also include those components and the gaseous pathways interconnecting them. In fact, in modern machines, the latter are often housed within the framework.

The vaporizers are mounted, either singly or in series, along the back bar, downstream from the flowmeter block. Traditionally, vaporizers were bolted onto the back bar and linked to each other by tapered fittings. The various manufacturers employed different sizes of tapers and mounting positions but these were superseded by the provisions of BS 3849 (UK), which recommended 23 mm 'cagemount' tapers. (The term cagemount originally refers to a type and size of tapered connection for a reservoir or rebreathing bag that has a small wire cage fitted to its inlet to prevent the neck of the bag from being obstructed,





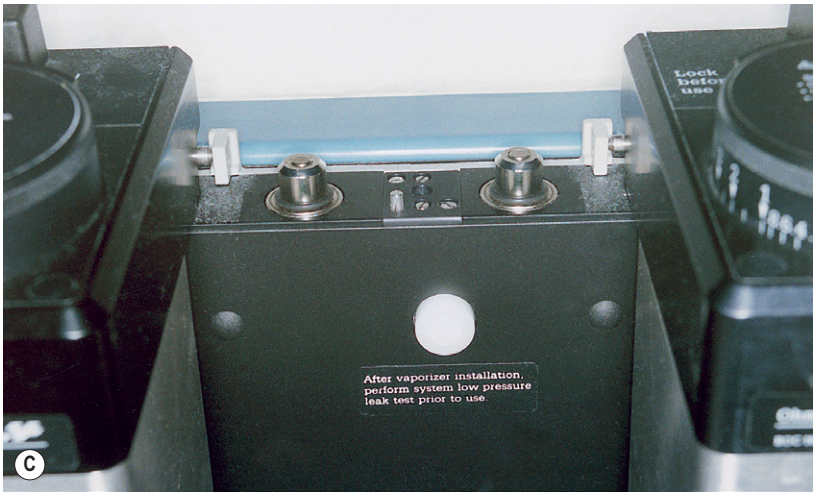


Figure 4.21 **A.** A three-station 'Selectatec' back bar (viewed from behind). **B.** Schematic diagram showing a TEC 5 vaporizer attached at station 1 and switched OFF, an empty station 2 and a TEC 5 attached at station 3 but switched ON. A, 'O'-ring seal; B, valve; C, TEC 3 lock-out pin; D, recess for vaporizer lock; E, safety interlock; F, recessed spindles in the vaporizer; G, immobilizer rod; H, back bar lock. **C.** A three-station back bar showing the safety interlock mechanism.

when the latter is empty and collapsed. The cagemount taper for vaporizers, though no longer used in the West, is still in use in many parts of the world.)

Modern vaporizers are designed so that they can easily be removed from the back bar and replaced by those for another agent. Systems in which vaporizers may be detached are generally regarded as an advance over the permanent cagemount system. Ease of removal has resulted in a greater flexibility in the choice and use of agents, and also ensures that anaesthetic machines do not have to be taken out of use to allow the servicing of the vaporizers.

Thus, the back bar provides mounting blocks as described below. The Ohmeda 'Selectatec' fitting is perhaps the most popular in the UK but most machine manufacturers now offer their machines with a choice of vaporizer mount.

The Ohmeda 'Selectatec' System

Each Selectatec station on the back bar has two vertically mounted male valve ports (Fig. 4.21). Between these inlet and outlet ports is an accessory pin and a locking recess. The matching vaporizer assembly has two female ports between which there is a locking assembly and a recess to accommodate the pin. The vaporizer is lowered on to the male valve ports and the locking knob is turned to fix it into the recess on the back bar (Fig. 4.21A). Successive generations of GE /Ohmeda vaporizers are labelled sequentially and, although the essential design of the back bar mounting system has not changed, the accessory pin has been added to prevent use of Tec 3 and previous generations on the modern back bar (see below).

'O'-rings on the male valve ports ensure a gas-tight fit. The two female ports on the vaporizer have recessed spindles (TEC 4 and above) that, when the vaporizer is switched on, protrude through the gas-tight seals of the male valve ports on the back bar. The ball valves (which provide the seals) in the male ports are displaced downwards occluding the back bar, and gas from the back bar is diverted into the vaporizer (Fig. 4.21B, station 3). TEC 3 vaporizers had fixed spindles that automatically depressed the ball valves in the male valve ports when the vaporizer was lowered on to the back bar assembly. Gas, therefore, passed through the head of the vaporizer even when it was not switched on or even locked on. This arrangement obviously had a greater potential for gas leaks and has been modified by the retractable spindle assembly on the TEC 4, 5, 6 and 7.

These models also incorporate a 'safety interlock'. This consists of an extension rod that protrudes sideways from a vaporizer as it is turned on, and displaces the equivalent rod on the vaporizer beside it, preventing the latter from being switched on. On a three-station back bar, there is a plastic lever (Fig. 4.21) linking stations 1 and 3. Should station 2 be empty, the lever links the extension rods between vaporizers 1 and 3 to ensure that only one vaporizer can be turned on at any one time. Tec 3 vaporizers had no safety interlock and this is yet another reason for their use being precluded by the fitting of the accessory pin. Presently machines are rarely specified with a three station back bar and this plastic lever is seldom required.

Dräger Interlock 2

The mounting system is similar to the 'Selectatec' version, although the dimensions are unique. The mounting system also includes an interlock preventing the administration of more than one vapour with a selector lever that locks one vaporizer at a time (Fig. 4.22). Dräger vaporizers are also available with Selectatec and other fittings.

Problems with detachable vaporizer systems

Removable vaporizer systems generate specific problems:

- As mentioned above, there is a greater potential for leaks.
- The vaporizer may be accidentally dropped and damaged in transit to or from the back bar.
- Tipping of older models of vaporizer in transit could result in liquid agent entering the bypass system causing either liquid or high concentrations of vapour to be present in the breathing system.
- Also, in countries where trichloroethylene is still used, a vaporizer containing this agent may accidentally be attached to a back bar station in a position that results in the vapour being passed into a breathing system containing soda lime. Trichloroethylene is known to react with warm soda lime to produce substances that are neurotoxic if inhaled.

Back bar working pressures

The flowmeter tubes in the flowmeter bank have, as a rule, been calibrated for gas flows assuming no downstream resistance. In a traditional back bar (23 mm internal diameter system) with the vaporizers switched off, the wide bore of the gas passages offers minimal flow resistance and so the back bar pressure developed at conventional flow rates (5–10 l min^{-1}) is marginally above atmospheric pressure. However, many modern back bars have narrow bore (8 mm) gas passages, which increase flow resistance and thus back-pressure on the flowmeters. The addition of high-resistance vaporizers and minute volume divider ventilators, which cause a build-up of pressure in the fresh gas flow (see Chapter 9, Automatic ventilators), increases back bar pressures. Table 4.1 shows typical back bar pressures developed and percentage changes in flowmeter settings with the 'Selectatec' back bar with and without a high-resistance vaporizer fitted. It should be noted that the small decreases in the flowmeter indications produced does not mean a decrease in the flow of gas to a patient. It is merely that the gas is compressed at the

higher pressures and subsequently re-expands downstream when the various resistances have been overcome. Readjustment of the flowmeters to the original settings following an induced pressure rise would therefore be inappropriate.



Figure 4.22 Dräger Interlock 2 system, and Vapor 2000 vaporizers.

Additional safety features

Several safety features are installed either on or downstream of the back bar:

- Intermittent back-pressure surges from certain minute volume divider ventilators can adversely affect vaporizer performance (see Chapter 3), and so most machines employ a spring-loaded non-return valve in the system to prevent these surges reaching the vaporizers.
- Since high-pressure build-up in the back bar can damage flowmeter and vaporizer components, a pressure relief valve (commonly set at 30–40 kPa) is fitted. This is often fitted in the same housing as the non-return valve (Fig. 4.23).

Emergency oxygen

A flowmeter bypass valve for an emergency oxygen supply is now fitted as standard (BS EN ISO 60601-2-13). The bypass flow joins the pipeline from the back bar just before the common gas outlet such that, when activated,

Table 4.1 Recorded gas pressures in a two-station Selectatec back bar

SITE OF PRESSURE MEASUREMENT	RECORDED PRESSURE (WITH NOMINAL FLOW RATES AT ATMOSPHERIC PRESSURE)		*PERCENTAGE CHANGE IN FLOWMETER SIGHT READINGS AT	
	5 l min^{-1}	10 l min^{-1}	5 l min^{-1}	10 l min^{-1}
Beginning of back bar (no vaporizers *in situ*) kPa (cm H_2O)	1.18 (12)	4.2 (43)	None	Minimal
At second vaporizer station (no vaporizers *in situ*) kPa (cm H_2O)	0.78 (8)	2.45 (25)	None	Minimal
Beginning of back bar with TEC 4 vaporizer at second station set to deliver:				
0% kPa (cm H_2O)	1.18 (12)	4.2 (43)	None	Minimal
1% kPa (cm H_2O)	3.23 (33)	8.5 (87)	None	<5%
5% kPa (cm H_2O)	2.74 (28)	7.74 (79)	None	<5%
Total occlusion of common gas outlet kPa (cm H_2O)	30.5 (312)		20%	20%

*This column shows the percentage change in sight readings, in a flowmeter initially calibrated at atmospheric pressure caused by the various resistances to flow seen in a 'Selectatec' back bar and TEC 4 vaporizer.

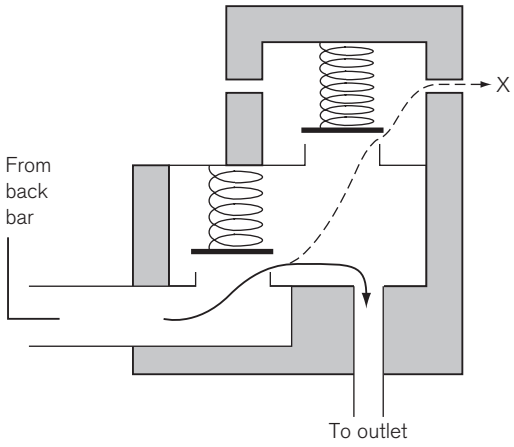


Figure 4.23 Combined non-return and pressure-relief valve at the end of the back bar. If the outlet is obstructed, the gasses escape at X, so protecting the back bar from overpressure. A low-pressure relief valve is also available to protect the patient.

it preferentially supplies oxygen at a rate of not less than 30 l min^{-1} into an attached breathing system. In earlier anaesthetic machines this bypass for oxygen was fitted near the flowmeter block. When it was operated, this resulted in an initial surge of gas and vapour to the patient prior to the pure oxygen being delivered.

The flowmeter bypass valve should no longer have a locking facility since this is regarded as dangerous and has resulted in cases of barotrauma when it has been switched on accidentally. The valve knob should also be recessed to minimize the chances of its inadvertent operation.

Oxygen failure warning devices

These were first introduced in the 1950s as a response to the problems of unobserved emptying of oxygen cylinders. However, early models could be unreliable as the battery-powered part of the alarm could be switched off or the battery could be exhausted or missing. The gas-powered part, which relied on nitrous oxide, could also be switched off or fail simultaneously with the oxygen (in which case the alarm would also not work).

The Ritchie whistle

The Ritchie whistle was introduced in the mid-1960s and forms the basis for most current oxygen failure devices. It was the first device to rely exclusively on the failing oxygen supply for its power. Figure 4.24 shows an oxygen failure warning device incorporating a Ritchie whistle marketed at one time by Ohmeda and still present on older machines in service.

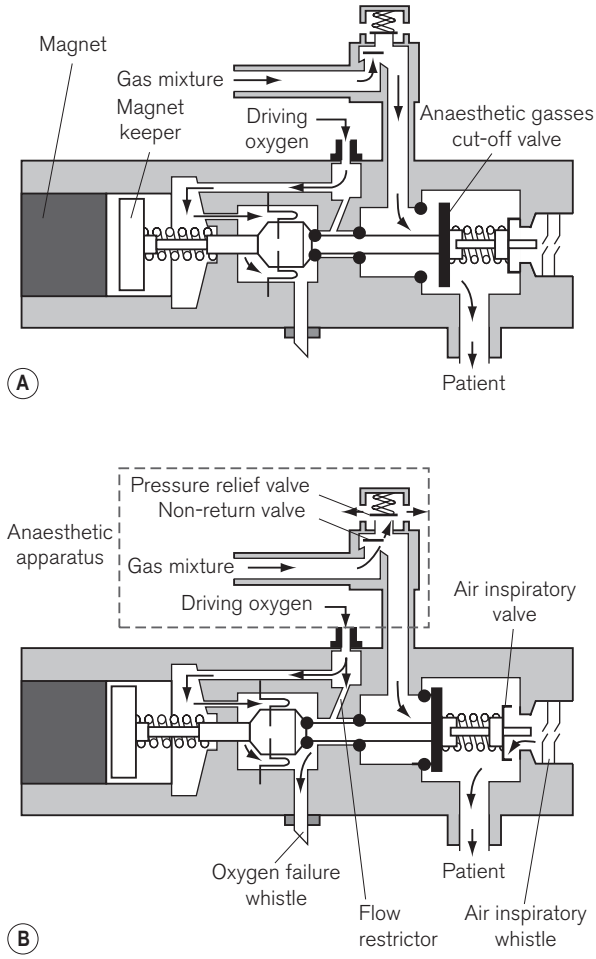


Figure 4.24 Oxygen failure warning device. **A.** Normal operation. **B.** Operation during oxygen failure.

The alarm is powered by an oxygen supply at a pressure of 420 kPa (60 psi) in the UK, which is tapped from the oxygen pipework upstream of the flowmeter block. This enters the alarm inlet valve and pressurizes the rolling diaphragm, opening the anaesthetic cut-off valve, and closing the air inspiratory valve and the port to the oxygen failure whistle. Anaesthetic gasses may then pass freely through this device, which is now at standby. The valve is kept in this position by the pressure of the oxygen supply opposing the force of the magnet and the return spring.

Decreasing pressure in the oxygen supply to the flowmeter block activates the valve, permitting a flow of oxygen (via the restrictor) to operate the oxygen failure whistle. The whistle sounds continuously until the oxygen pressure has fallen to approximately 40.5 kPa (6 psi). At a pressure of approximately 200 kPa (30 psi) the force of the magnet keeper return spring and the magnet causes the anaesthetic

gasses cut-off valve to be closed, cutting off the supply of anaesthetic gasses to the patient. At the same time the spring load on the air inspiratory valve is released, allowing the patient to inspire room air. Whenever the patient inhales, the inspiratory air whistle sounds.

With the anaesthetic gasses cut-off valve closed, the now potentially hypoxic gas from the flowmeter block vents to the atmosphere through the pressure-relief valve on the back bar.

Current oxygen failure warning devices

BS EN 740:1999 and EN ISO 60601-2-13:2006, its replacement, stipulate that anaesthetic workstations in use with any gasses containing less than 21% premixed oxygen content (e.g. pure nitrous oxide or carbon dioxide) shall be operated with a gas cut-off device. This gas cut-off device must either:

- cut off the supply of all gasses other than oxygen, air and premixed gasses with an oxygen content above 21%(V/V) to the fresh gas outlet; or
- progressively reduce the flow of all other gasses (except air or premixed gasses with an oxygen content above 21% (V/V)) while maintaining the proportion of oxygen until the supply of oxygen finally fails, at which point the supply of all other gasses (except air or premixed gasses with an oxygen content above 21% (V/V)) shall be cut off.

Also 'the gas cut off device shall not be activated before the oxygen supply failure alarm', and 'the sole means of resetting the gas cut-off device shall be restoration of the oxygen supply pressure to a level above that at which the device is activated'.

A gas-powered auditory alarm signal for oxygen failure is required to be of at least 7 s duration and where the alarm signal is gas powered (as opposed to electrically generated) 'the energy required to operate it shall be derived from the oxygen supply pressure'.

Figure 4.25 shows a schematic diagram of the pneumatic arrangement for an alarm and shut off system that satisfies current standards.

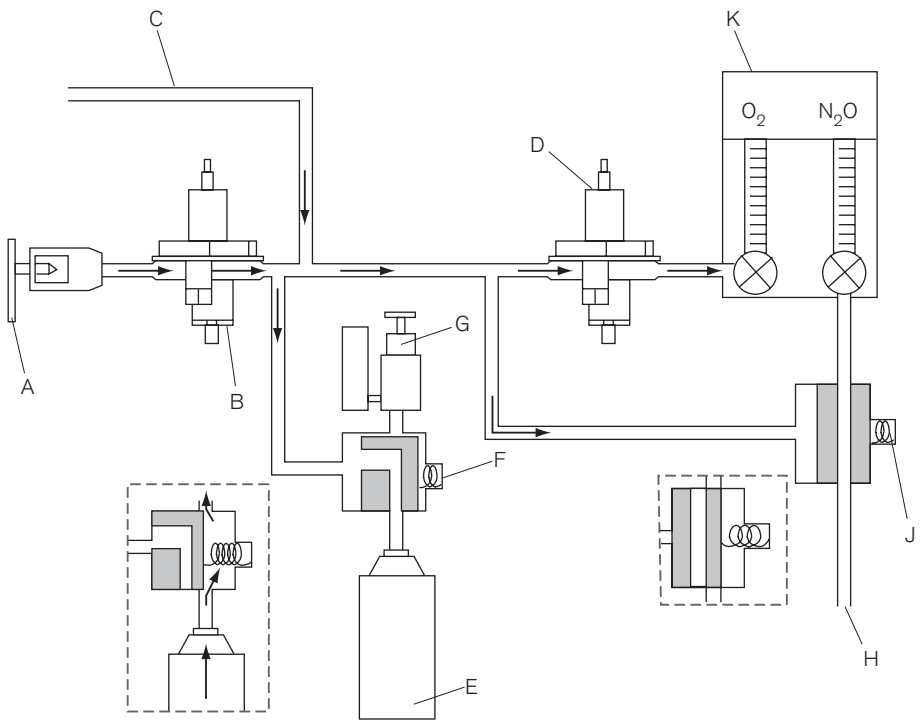


Figure 4.25 A schematic diagram of a current oxygen failure warning device. A, Cylinder yoke for oxygen; B, primary regulator for oxygen (13700 kPa to 420 kPa); C, pipeline oxygen supply; D, secondary regulator for oxygen (420 kPa to 140 kPa); E, reservoir of oxygen required to power the Ritchie whistle for a minimum of 7 s; F, spring-loaded regulator. When oxygen supply pressure drops to 200 kPa reservoir E is connected to the Ritchie whistle; G, Ritchie whistle; H, nitrous oxide supply; J, spring-loaded shut-off valve to nitrous oxide supply activated when oxygen supply pressure drops below 200 kPa; K, flowmeter bank.

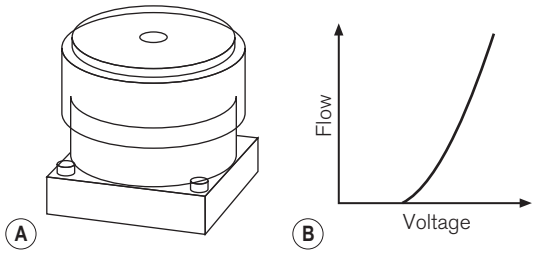


Figure 4.27 **A.** Representation of proportional flow valve and **B.** the relationship between voltage and gas flow. (Fig. 9.8A, shows the mode of operation of a typical proportional flow valve.)

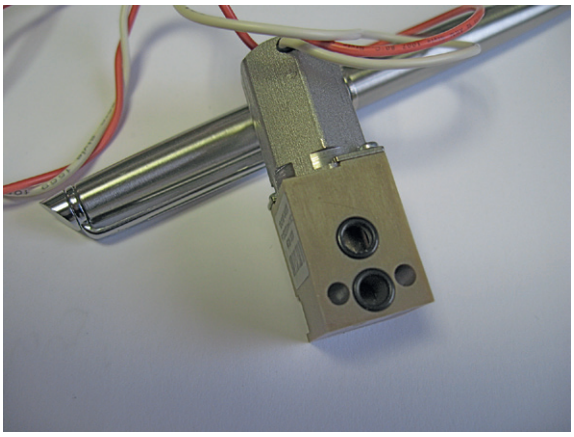


Figure 4.28 Proportional valve (resting on an ordinary writing pen for scale) as used in the Blease900 Series ventilator (Spacelabs Healthcare, USA).

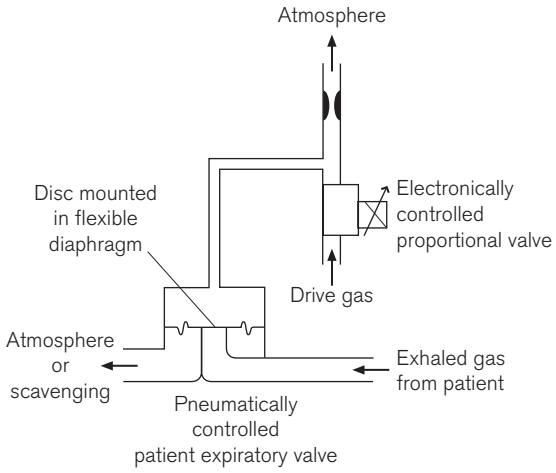


Figure 4.29 Proportional valve control of patient expiratory valve.

with a maximal response time of less than 100 ms. In ventilators the valve is usually placed immediately downstream of a secondary regulator which supplies the drive gas to the valve at a pressure of about 2.3 bar. Manufacturers set this pressure somewhat below the minimum pressure that may be expected from gas pipelines in order to have a non-fluctuating supply pressure to the valve so that pressure/flow characteristics are predictable for the flow controller algorithm. ITU ventilators may have larger, more sophisticated proportional valves with an order of magnitude swifter response times. Massive flows can be rapidly generated on patient demand, even allowing almost seamless pressure support of spontaneous patient breathing. Downstream flow and pressure sensing permits feedback control of the flow from the valve via the microprocessor control system.

Control of such valves is accurate enough to permit positive end-expiratory pressure (PEEP) to be applied in some ventilators simply by allowing a bias flow through the valve during expiration. Sufficient flow, controlled by a feedback loop, is allowed to pressurize the bellows and expiratory (pop off) valve in the bellows housing to give the desired level of PEEP.

Some systems, such as the Dräger E series ventilators, use a separate or second proportional valve to control a lower pressure gas flow from which a side arm is used to pressurize the machine side of an expiratory valve for control of inspiration and PEEP or continuous positive airway pressure (CPAP) (Fig. 4.29). Alternatively this proportional valve can be of a type that operates an actuator directly onto the leaflet of the ventilator inspiratory/expiratory control valve (Fig 4.30). (See also Figs 9.8 and 9.9, solenoid driven proportional flow valve for low-pressure gas.) Note, comparing figures 4.28 and 4.30, that the valves that manage low pressure gas flows have very much larger apertures.

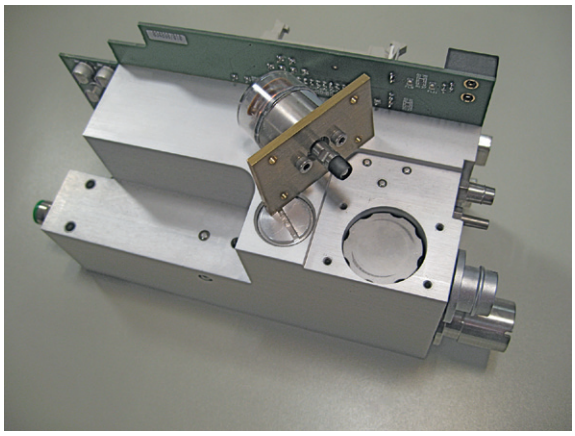


Figure 4.30 Blease900 series proportional valve directly acting on a valve leaflet for inspiratory/expiratory phase control.

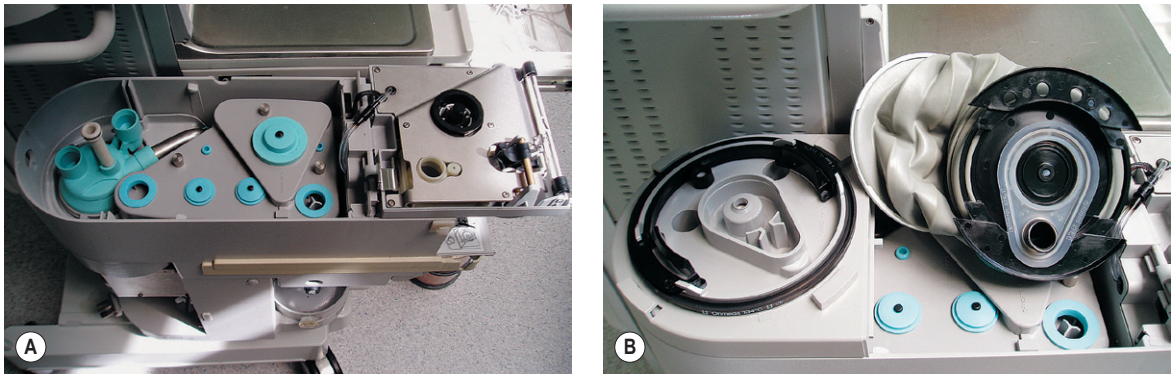


Figure 4.31 **A.** Circle system of Datex-Ohmeda Aestiva partially stripped down to show gas pathways, bellows, valves, seals and other autoclaveable parts (including blue silicone rubber parts and 'airbox' in centre of upper photograph). **B.** Shows the expiratory pop off valve at the base of the bellows.

INTEGRAL BREATHING SYSTEMS

Concerns about economy in the use of volatile anaesthetic agents and gasses, for reasons of both cost and atmospheric pollution, coupled with the heat and moisture retaining properties of the circle system, have ensured that the circle has become an integral part of the modern anaesthetic workstation.

The need to both integrate the breathing system into an ergonomic design and allow sterilization of patient gas pathways, can produce a labarynthine system with multiple parts, valves and microswitches to sense the position and function of levers and detachable components (Fig. 4.31). That in spite of such numerous parts, many of these machines cannot be misassembled and do not leak, is a testament to good engineering design. However, the potential for problems to arise in such complex devices must always be borne in mind and it is advocated that a self-inflating bag and alternative source of oxygen is always readily available (see Pre-use check below). A 'top circuit', such as a Mapleson C or D for attachment to the common gas outlet in times of crisis and confusion, is also extremely useful.

The water generated by the circle absorber system is beneficial in moisturizing the inspired patient gasses, but poses a problem for designers of anaesthesia machines particularly with the increasing use of low flow anaesthesia. *Rain out* or condensation of this moisture in the system can interfere with the function of the valves and flow-metering within these systems. Particular provision must be made for this, and the issue is tackled differently in the various machines by the manufacturers. This may be the use of heated patient circuit plates as in the Dräger Primus

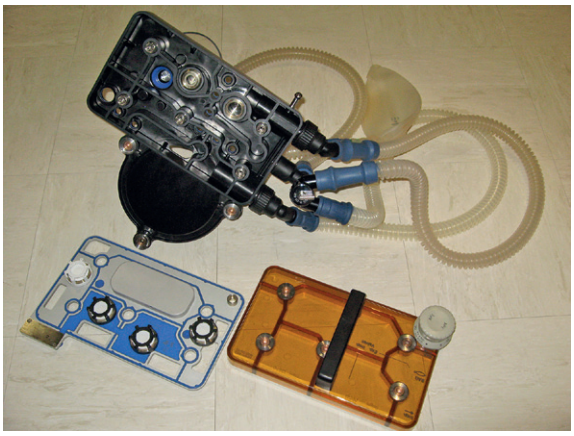


Figure 4.32 Dräger Primus patient breathing system disassembled to show heated steel plate in foreground on left.



Figure 4.33 Condenser fitted to GE Healthcare Aisys. See text.

(Fig 4.32), or the use of an additional specific condenser arrangement as in the GE Healthcare Aisys (Fig. 4.33). Here this approach is coupled with flowmeters on the circle system that are modified so that water accumulation is less likely to interfere with the function of the flap valve of the differential pressure variable orifice flowmeters. In



Figure 4.34 Circle system drain plug in Aestiva/5, GE Healthcare. See text.

the Aestiva also by GE Healthcare the point of condensate collection is at the bottom of the absorber assembly and a drain plug is placed there (Fig. 4.34). This is a frequent source of a breathing system leak that is difficult to trace as it is not obvious when this drain plug is open.

ERGONOMICS

Ergonomics is the study of the efficiency of persons in their working environment and *human factors engineering* is the design and development of equipment to improve the ergonomics of a task. This has the effect of making the working environment not only more pleasant, but also less tiring and less stressful, which should lead to increased safety. The conventional pneumatic anaesthetic machine 'just grew' and there has been little attempt in the past to specifically consider ergonomics. Nonetheless, some design endpoints, reached after a long journey of reducing error and set in current standards, such as those specifying the stand out position and shape of the oxygen flowmeter control knob (together with attendant guards to prevent accidental resetting), exemplify ergonomics in practice.

is a time delay before a problem is noticed by the anaesthetist and the cause identified. Another delay occurs before the problem is corrected, after which there should be a recovery to safe conditions if the correction is made early enough. Excessive delay in noticing the problem or in its correction may lead to permanent injury.

The reliability of the human for constant vigilance over long periods of time is questionable and the ability to make decisions when bombarded with multiple sensory inputs is sorely put to the test. Alarm systems should be designed to allow for as much time as possible to correct a problem before injury begins. Examination of Figure 4.36 shows that this may be accomplished by minimizing the pre-alarm period and making it clear to the anaesthetist what the problem is and its level of urgency. Research has shown that *intelligent alarm systems* which integrate and prioritize multiple alarm conditions can lead to a more rapid and consistent rectification of adverse incidents.[5]

The further along the system (e.g. from the gas and electricity supplies to the patient) that a parameter is monitored (Fig. 4.37), the greater the delay before the alarm is sounded and the greater the number of possible causes of that particular problem. For example, if the oxygen supply fails, the oxygen supply alarm would sound immediately. Some seconds later, depending on the fresh gas flow into the breathing attachment, the inspired oxygen monitor alarm would become active. However, it may be more than a minute before the saturation, as indicated by a pulse oximeter, would fall below the critical level. Furthermore, the causes of a drop in SpO_2 are numerous compared with the causes of the sounding of the oxygen supply pressure alarm. In this scenario, assuming multiple monitoring points of anaesthesia system and patient, an intelligent alarm system would assimilate the multiple alarm conditions and prioritize the oxygen supply failure, hence leading to the most rapid resolution of the incident.

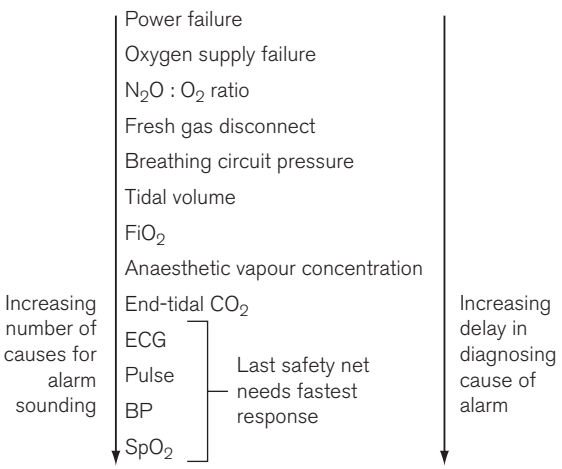


Figure 4.37 The further along the system from the supplies to the patient that a parameter is monitored, the greater the delay before the problem is corrected.

Alarms

Alarms do not necessarily refer to emergencies, but may indicate abnormal situations that may or may not have the potential to become emergencies. Given the number of different parameters surveyed by the workstation monitors, there clearly has to be some system of prioritizing alarms to allow sensible signalling in multiple alarm conditions. Alarm conditions are, therefore, given a hierarchy from *advisory* (requiring awareness) to *caution* (requiring a prompt response) and *warning* (requiring immediate response). At any time, the condition with the highest priority is preferentially displayed. Ideally an audible warning differentiating between these three levels draws the anaesthetist's attention to a visual indication of what the problem is. *Latching* of alarms means that even if the condition leading to the alarm is resolved, the alarm continues to sound until it is acknowledged and reset. Accepted standards give minimum alarm hierarchies for various parameters and state those that can and cannot be disabled.

Much research has been done on alarms,[6] alarm characteristics and those features that control the perceived urgency, namely: frequency composition, repetition rate, amplitude, and harmonic relation of the frequency components. The urgency of the alarm must be balanced against its liability to distract the target's concentration during critical periods, and the possible nuisance effect from 'false alarms', as well as its effect on non-target audience (e.g. patient or surgeon). This is not an easy task; additionally, an alarm perceived as excessively intrusive may be disabled by the anaesthetist. Although the BSI and ISO publish clear standards for alarms (BS EN ISO 9703–3:1998), significant latitude is still allowed to manufacturers and even with standards compatible alarm structures it is argued that the current alarms in many popular machines are inappropriate in terms of conveying the correct urgency.[7] The 9703 series of standards was withdrawn in 2004. Ideally, however, ultimately alarm signals should be identical across manufacturers for any given modality of monitoring or item of equipment. In any case, it is perhaps time manufacturers revisited this subject.

Nonetheless, the quality of modern monitoring equipment available, as well as the ease and flexibility of operator setting of alarm limits, means that the practice of disabling alarms (rather than setting alarm parameters to prevent inappropriate alarm triggering) is no longer acceptable. By the same token, alarms should not be allowed to continually sound (often a feature of inadequate implied urgency) when there is no perceived problem: this leads to

habituation to the alarm sound and ultimately failure to react to appropriate alarms.

PRE-USE CHECK

It is mandatory to check the correct functioning of anaesthetic equipment before use. The Association of Anaesthetists of Great Britain and Ireland (AAGBI) publishes a 'Checklist for Anaesthetic Equipment' to assist in allowing a comprehensive and systematic check of equipment (Table 4.2). The most recent iteration of this, in 2004,[8] recognizes the introduction of microprocessor controlled technology into anaesthetic workstations and attempts not to reduplicate the self-test cycle that modern workstations undergo on start-up. It also stresses the importance of familiarity with these potentially complex pieces of machinery and hence the need for a formal 'induction' onto the machine. Also noteworthy is the requirement that a log should be kept of the completion of a pre-use check according to both the requirements of the manufacturer and the AAGBI checklist.

With all machines and, perhaps, even more so with electronically controlled workstations, consideration must be given to the possibility that the machine will fail to deliver any or adequate flows of gas. For this reason it is imperative to check that an alternative oxygen supply and means of ventilation, such as a self-inflating bag, are readily available and functional.

An abbreviated description of the AAGBI checking procedure is produced as a laminated sheet for attachment to each anaesthetic machine. This is reproduced below.

Table 4.2 AAGBI checklist for anaesthetic equipment (with permission from AAGBI)

THE ASSSOCIATION OF ANAESTHETISTS OF GREAT BRITAIN AND IRELAND CHECKLIST FOR ANAESTHETIC EQUIPMENT 2004

The following checks should be made prior to each operating session.

In addition, checks 2, 6 and 9 (Monitoring, Breathing System and Ancillary Equipment) should be made prior to each new patient during a session.

1. Check that the anaesthetic machine is connected to the electricity supply (if appropriate) and switched on.
   Note: Some anaesthetic workstations may enter an integral self-test programme when switched on; those functions tested by such a programme need not be retested.
   - Take note of any information or labelling on the anaesthetic machine referring to the current status of the machine. Particular attention should be paid to recent servicing. Servicing labels should be fixed in the service logbook.
2. Check that all monitoring devices, in particular the oxygen analyzer, pulse oximeter and capnograph, are functioning and have appropriate alarm limits.
   - Check that gas sampling lines are properly attached and free of obstructions.
   - Check that an appropriate frequency of recording non-invasive blood pressure is selected.

   (Some monitors need to be in stand-by mode to avoid unnecessary alarms before being connected to the patient.)
3. Check with a 'tug test' that each pipeline is correctly inserted into the appropriate gas supply terminal. Note: Carbon dioxide cylinders should not be present on the anaesthetic machine unless requested by the anaesthetist. A blanking plug should be fitted to any empty cylinder yoke.
   - Check that the anaesthetic machine is connected to a supply of oxygen and that an adequate supply of oxygen is available from a reserve oxygen cylinder.
   - Check that adequate supplies of other gasses (nitrous oxide, air) are available and connected as appropriate.
   - Check that all pipeline pressure gauges in use on the anaesthetic machine indicate 400–500 kPa.
4. Check the operation of flowmeters (where fitted).
   - Check that each flow valve operates smoothly and that the bobbin moves freely throughout its range.
   - Check the anti-hypoxia device is working correctly.
   - Check the operation of the emergency oxygen bypass control.
5. Check the vaporizer(s):
   - Check that each vaporizer is adequately, but not over, filled.
   - Check that each vaporizer is correctly seated on the back bar and not tilted.
   - Check the vaporizer for leaks (with vaporizer on and off) by temporarily occluding the common gas outlet.
   - Turn the vaporizer(s) off when checks are completed.
   - Repeat the leak test immediately after changing any vaporizer.

Continued

Table 4.2 AAGBI checklist for anaesthetic equipment (with permission from AAGBI)—cont'd

THE ASSSOCIATION OF ANAESTHETISTS OF GREAT BRITAIN AND IRELAND
CHECKLIST FOR ANAESTHETIC EQUIPMENT 2004

6. Check the breathing system to be employed.
   Note: A new single-use bacterial/viral filter and angle-piece/catheter mount must be used for each patient. Packaging should not be removed until point of use.
   - Inspect the system for correct configuration. All connections should be secured by 'push and twist'.
   - Perform a pressure leak test on the breathing system by occluding the patient-end and compressing the reservoir bag. Bain-type co-axial systems should have the inner tube compressed for the leak test.
   - Check the correct operation of all valves, including unidirectional valves within a circle, and all exhaust valves.
   - Check for patency and flow of gas through the whole breathing system including the filter and anglepiece/catheter mount.
7. Check that the ventilator is configured appropriately for its intended use.
   - Check that the ventilator tubing is correctly configured and securely attached.
   - Set the controls for use and ensure that an adequate pressure is generated during the inspiratory phase.
   - Check the pressure relief valve functions.
   - Check that the disconnect alarms function correctly.
   - Ensure that an alternative means to ventilate the patient's lungs is available (see item 10, below).
8. Check that the anaesthetic gas scavenging system is switched on and is functioning correctly.
   - Check that the tubing is attached to the appropriate exhaust port of the breathing system, ventilator or workstation.
9. Check that all ancillary equipment which may be needed is present and working.
   - This includes laryngoscopes, intubation aids, intubation forceps, bougies, etc., and appropriately sized facemasks, airways, tracheal tubes and connectors, which must be checked for patency.
   - Check that the suction apparatus is functioning and that all connectors are secure.
   - Check that the patient trolley, bed or operating table can be rapidly tilted head down.
10. Check that an alternative means to ventilate the patient is immediately available (e.g. self-inflating bag and oxygen cylinder).
    - Check that the self-inflating bag and cylinder of oxygen are functioning correctly and the cylinder contains an adequate supply of oxygen.
11. Recording.
    - Sign and date the logbook kept with the anaesthetic machine to confirm the machine has been checked.
    - Record on each patient's anaesthetic chart that the anaesthetic machine, breathing system and monitoring has been checked.

This checklist is an abbreviated version of the Association of Anaesthetists publication 'Checking Anaesthetic Equipment 3 2004'(endorsed by the Chief Medical Officer and the Royal College of Anaesthetists).

SPECIFIC MACHINES

By considering some additional aspects of just a few specific machines, it is possible to achieve a perspective on the scope of design of the modern workstations currently available. Also, an understanding of the workings of any one machine potentially gives insight into significant issues and features of these devices in general. Five machines are therefore considered separately below. When compared with each other it is interesting to observe their differing use of technology and the disparate focus of their respective designers. It will be interesting to see, over the next few years, which approach predominates and where the application of technology will prove most effective. Certainly, electronic control of the machine does not yet dominate, although it appears to be gaining more widespread acceptance than 5 years ago.

A number of manufacturers still produce simple pneumatic machines with no electronic components or integral breathing systems and there is clearly a role for such equipment. They are reliable, need minimal maintenance and are low cost and admittedly simpler to operate. They predominate in areas where non-anaesthetists may use them for resuscitation and in areas where space may be at a premium or other constraints operate, such as anaesthetic induction areas. Their design conforms to the principles outlined above under 'The anaesthetic delivery system'; certainly in the UK there should no longer be machines available anywhere capable of delivering an hypoxic mixture.

The following paragraphs are not intended as a critique or assessment of the machines. Features and failures are highlighted purely to make the reader aware of issues that may be pertinent to any device being considered. The purpose of anaesthetic machines is ultimately quite simple – the safe delivery of drugs. Increasing familiarity with a machine can therefore only reveal its inadequacies.

(Datex-Ohmeda) Aestiva/5, GE Healthcare

The Aestiva/5, and its family of machines launched in 1998, has been a very successful design of anaesthetic workstation (Fig. 4.38). It is, in all respects, a traditional pneumatic machine. There is a built-in circle system and a software-driven, pneumatically powered ventilator using the 'enclosed bellows' arrangement to isolate respiratory and driving gasses. The ventilator type is briefly described above and specifically in detail in Chapter 9 (Datex-Ohmeda 7900). Conventional vaporizers are mounted on the back bar. Hypoxic mixture prevention is by use of the Ohmeda Link 25 system (see Fig. 4.17). The CGO, as opposed to the integral circle breathing system, is selected by a lever discretely positioned to the side of the absorber.

Its popularity owes more to good ergonomic design than the use of advanced technology. Without doubt,

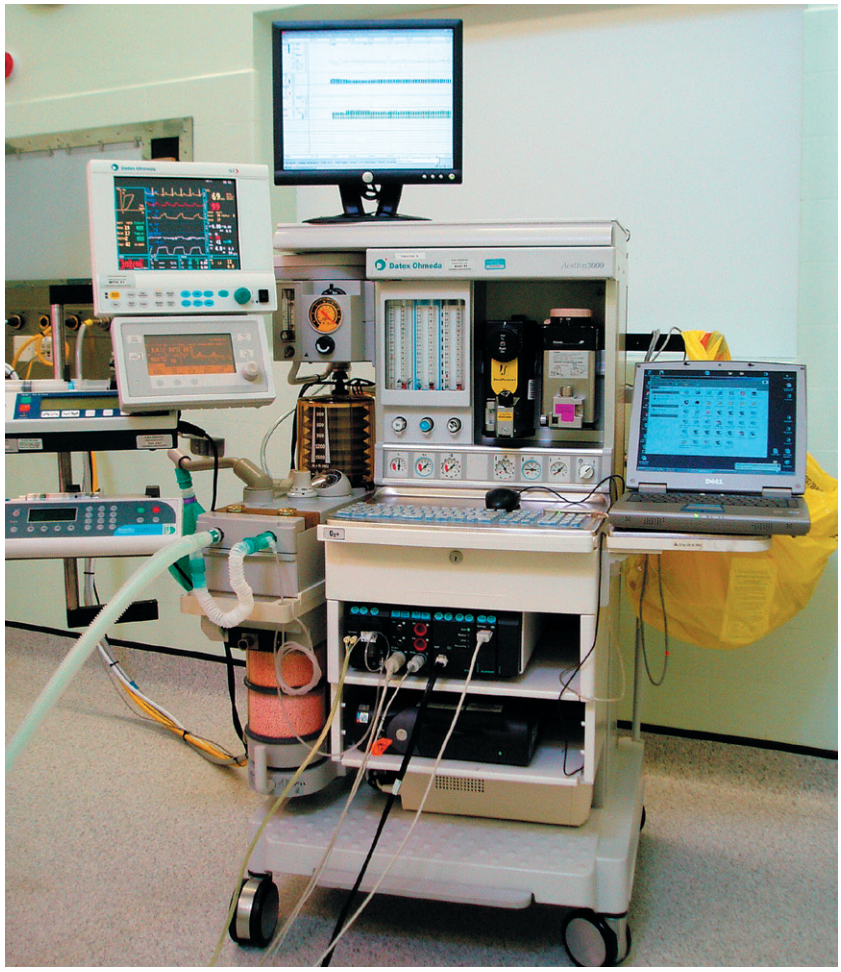


Figure 4.38 The Datex–Ohmeda Aestiva/5 workstation with S/5 monitor in use. A networked record keeping system and a further networked computer are also seen.

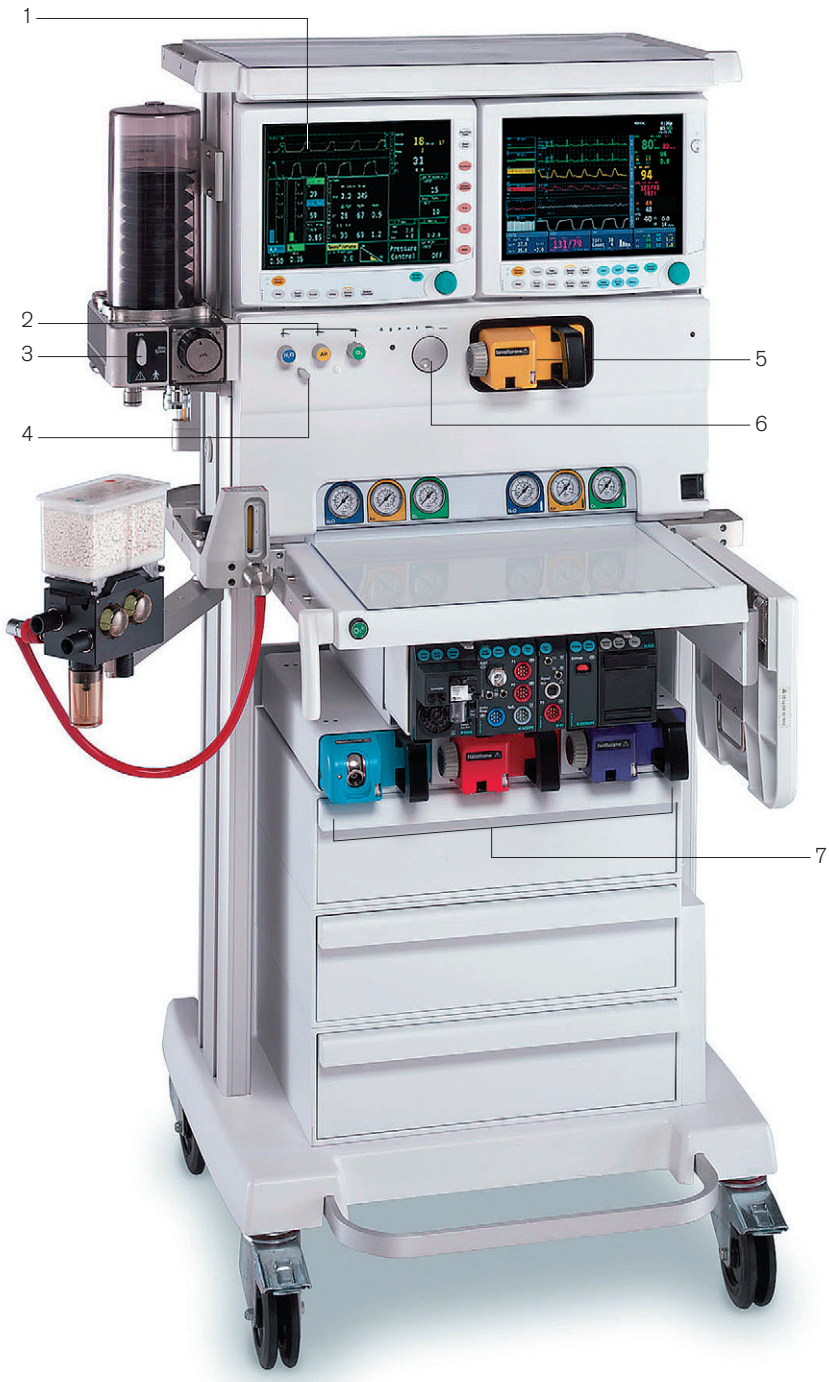


Figure 4.39 The Datex-Ohmeda ADU. Shown here with US standard colour coding for gasses. (1) Ventilator user interface. (2) Gas flow needle valve controls. (3) Ventilator selection switch. (4) Side gas selection switch. (5) Aladin cassette. (6) Agent concentration selection. (7) Vaporizer storage rack. Photograph courtesy of Datex-Ohmeda.

much of this is due to the long-established Datex-Ohmeda AS/3 design of patient monitor. This monitor has hot swappable individual modules for an extensive number of parameters and benefits from a flexible and well laid out display and intuitive operating menu structure. This monitor is neatly housed within the Aestiva with a remote flat panel display which can be mounted on the swing arm of the machine to sit above or below the control panel of the ventilator. Depending on the configuration requested, the machine can offer a significant amount of flat storage or working surfaces and drawers. Use of a single, foot-operated brake lever renders the machine freely mobile, allowing easy positioning of machine for access to the patient in any operating room set-up. The overall appearance of the machine is unfussy with clearly defined instrument clusters.

Those new to the machine should be aware that the low-urgency ventilator panel alarms, such as the *advisory* alarm for selection of the auxiliary common gas outlet, are too easily ignored. One area where the machine would benefit from a more intelligent alarm structure is in the linking of respiration detected on the sensors of the integral circle system to the position of the auxiliary common gas outlet selection lever. Currently, with no increase in alarm urgency, a patient can breathe spontaneously on the circle system with volumes monitored at the circle displayed on the ventilator panel, whilst anaesthetic gasses are delivered elsewhere.

(Datex-Ohmeda) ADU, GE Healthcare

First released in 1995 ('ADU 95') and with a subsequent 'ADU 98', this is still a striking-looking machine by virtue of the absence of traditional rotameters and vaporizers and the presence of twin flat panel displays for patient monitoring and control and monitoring of the pneumatic systems (Fig. 4.39). GE Healthcare (who took over Datex-Ohmeda) ceased production of this device in February 2010 with over 11 000 units having been put into service worldwide.

The ADU is essentially a modular system and comprises:

- a wall gas unit, taking gas inlets from pipeline supplies and reserve cylinders
- a fresh gas control unit responsible also for volatile agent admixture
- an electronic ventilator unit
- a central electronic unit for integrating control and monitoring
- an attached bellows block and compact circle absorber system.

The ventilator and fresh gas control units have their own microprocessor controls which communicate through the central electronic unit. A number of features of the ADU stand out for individual consideration (Fig. 4.40).

In spite of the absence of traditional flowmeters, gas flow is set by the operator using standard needle valves. An electronic switch selects air or N_2O as the 'side gas'. Electronic flow measurement across a laminar flow restrictor for both oxygen and the side gas ensures a minimum 25% oxygen concentration by controlling a proportional valve downstream of the N_2O needle valve when N_2O is selected. Gas flows are shown as pictograms on the user interface which also has a comm wheel for selecting and setting ventilator parameters.

The ventilator is a sophisticated microprocessor controlled pneumatically driven (air or oxygen) design employing a number of proportional valves for individual control of inspiratory flow, expiratory valve and PEEP valve. Pressure and flow are measured at a number of points in the pneumatic system to allow close control of respiration. A single lever on the bellows block selects between bellows or reservoir bag and APL valve and relays via a sensing microswitch to switch on the ventilator.

At the time of its inception it was unique in its electronic control of agent concentration. The Aladin cassette system none the less relies on the conventional approach of saturation of a proportion of the fresh gas flow by passage through a vaporizing chamber. Here though, when the desired agent concentration is selected on the machine, a proportional valve controls the flow of gas through the vaporizer. By comparing the bypass flow measurement within the machine with the vaporizer flow measurement, the proportional valve is set to achieve the desired target (Fig. 4.40). The system is described in greater detail in Chapter 3. Note that in the ADU feedback control of the agent concentration relies on calculations of flow and does not involve measurement of agent concentration.

The patient monitoring unit of the ADU will be familiar to many as the Datex-Ohmeda AS/3 or S/5 and is a separate modular monitor with no linkage to the control unit of the ADU.

Dräger Primus

The Dräger Primus, available since 2003 (Fig. 4.41), represented another departure in design. Its main distinguishing feature is the electronic fresh gas mixer.

Here, fresh gas is mixed entirely by electronically controlled valves according to the desired total flow and oxygen concentration. The ADU, for comparison, utilizes mechanical needle valves set by the operator with subsequent electronic adjustment to the flow of N_2O. In the Primus the operator selects air or N_2O as the carrier gas, and sets a total flow and the desired oxygen concentration (Fig. 4.42). A gas controller unit under feedback control sequentially opens high-speed solenoid valves to



Figure 4.41 The Dräger Primus. Photo courtesy of Dräger Medical UK.

anaesthetic machine (Fig. 4.45). Ergonomics is addressed by a central brake pedal, 180° rotating capability of the display panel arm and push through drawers allowing the machine to be used from either side. The outstanding features of the device are its automated control of patient gas composition using a choice of two different control algorithms and its use of a novel miniature turbine within the breathing system to power patient gas circulation and ventilation. The Zeus includes the following components:

- Windows NT computer with system display unit for operation of therapy control and monitoring
- Electronic gas and anaesthetic agent flow control with closed-loop feedback control system
- Electronically controlled and driven blower with circle breathing system
- Airway and gas monitoring
- Optional syringe pumps for intravenous anaesthesia (currently, however, TCI is not available on these pumps).

Like the Aisys below, the only direct mechanical controls are the emergency oxygen flush button (here duplicated on the opposite side of the machine), a deployable back-up oxygen needle valve control, and an additional O_2 flowmeter for supplemental oxygen. These sections of the machine do not need an electrical supply to be operational. The machine is otherwise entirely controlled through a touch-sensitive flat panel display.

Gasses are delivered to a digital gas mixer from secondary regulators at 2.4 bar. Gas mixing for the fresh gas flow into the patient circuit is then achieved by two banks of five digital switching valves (solenoids): one bank for control of oxygen and the other used for either air or

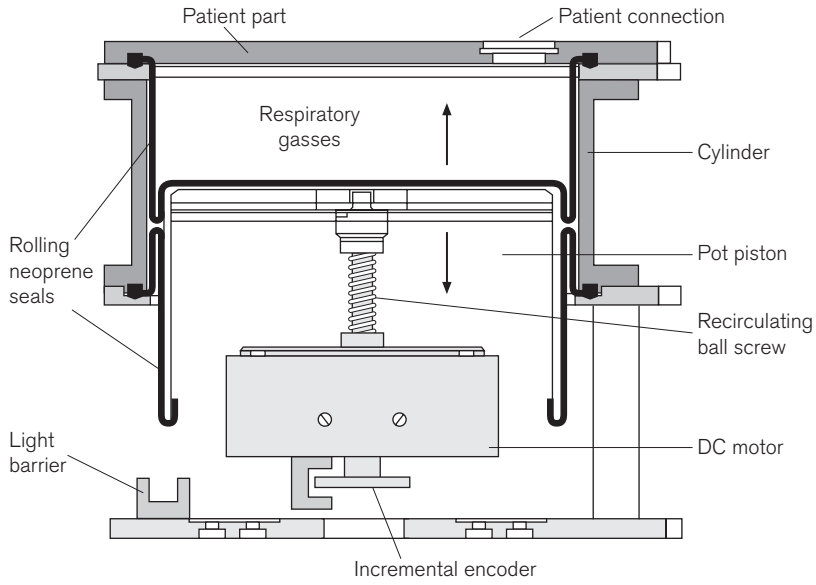


Figure 4.43 Schematic of Dräger E series ventilator.

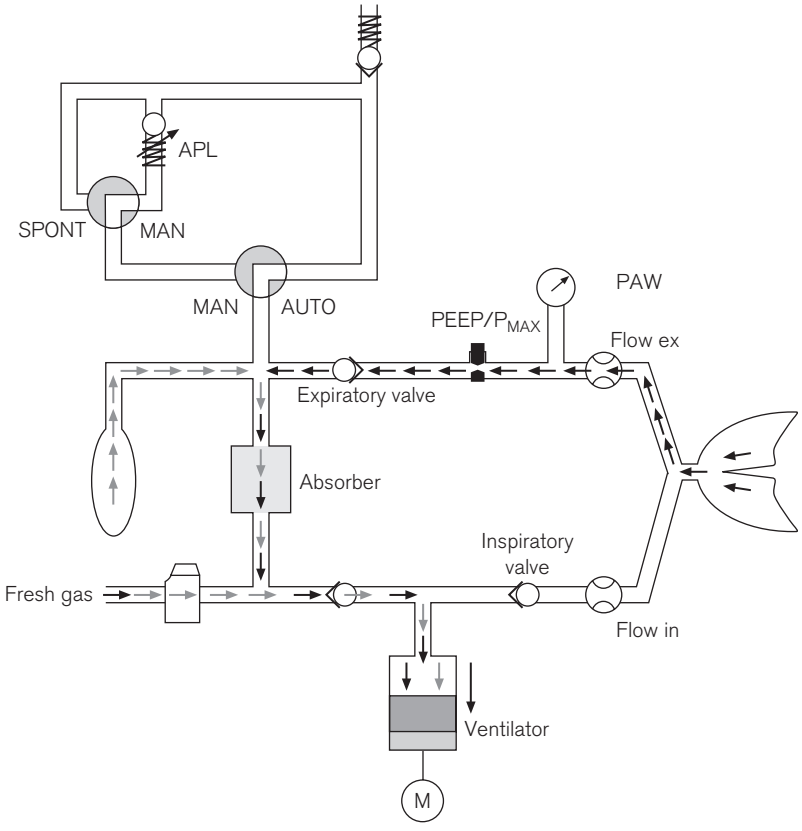


Figure 4.44 Gas flow in the Primus during expiration.

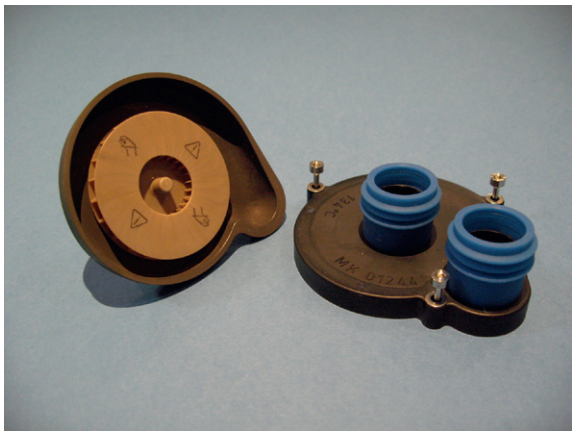


Figure 4.46 The turbine (left) powering ventilation in the Zeus (Dräger Medical, Lubeck, Germany) next to its patient circuit connection.

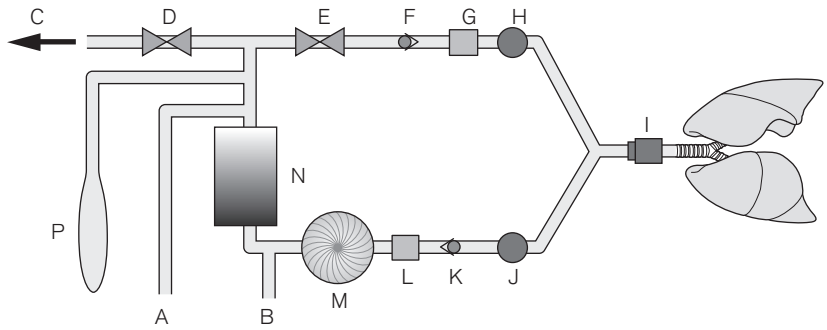


Figure 4.47 Zeus gas flow schematic. A, inlet for mixed fresh gasses; B, alternative inlet for fresh gas in the event of blower failure; C, feed into anaesthetic gas scavenging system; D, APL valve; E, Flow valve; F, non-return valve; G, expiratory pressure sensor; H, expiratory flow sensor; I, 'Y' piece gas sampling port; J, inspiratory flow sensor; K, non-return valve; L, inspiratory pressure sensor, M, blower; N, absorber; P, reservoir bag.

GE Healthcare, Aisys

The Aisys (coined form *A*naesthesia *I*ntegrated *Sy*stem) launched in the UK in 2006 is this manufacturer's fully electronically controlled anaesthetic workstation (Fig. 4.1). In contrast to the Datex-Ohmeda ADU (see above), this model uses electronic control of gas flow into the patient circuit, doing away with the needle valves seen in the ADU. It is in all other respects a fairly conventional design. The vaporizer module uses identical technology to the ADU and the ventilator assembly is the conventional proportional valve driven enclosed bellows system of the Datex-Ohmeda 7900 series. Unlike the externally connected breathing system of the ADU, however, the circle breathing system here is very much integral to the machine. Gas flow into this circle system is such that for use with other breathing systems (e.g. Mapleson A or D), by selecting a non-circle mode on the machine control panel the inspiratory outlet of the 'circle' may be used as a common gas outlet (CGO). The machine may also be optioned with a switchable separate CGO as in the Aestiva.

Fig. 4.48 shows the pneumatic system of the Aisys. The emergency oxygen flush button and the auxiliary O_2 flowmeter are direct mechanical systems and do not require the machine to be powered on. The alternate O_2 flowmeter is in series with the master on/off switch and is provided as a parallel back-up system in case of failure of the display unit or electronic control of the gas mixer when the workstation is switched on and functional (Fig. 4.49). Under such circumstances it should deploy automatically and allow the manual control of oxygen into the patient circuit. Depending on the source of the problem, it is possible that electronic control of gas flow to the vaporizer

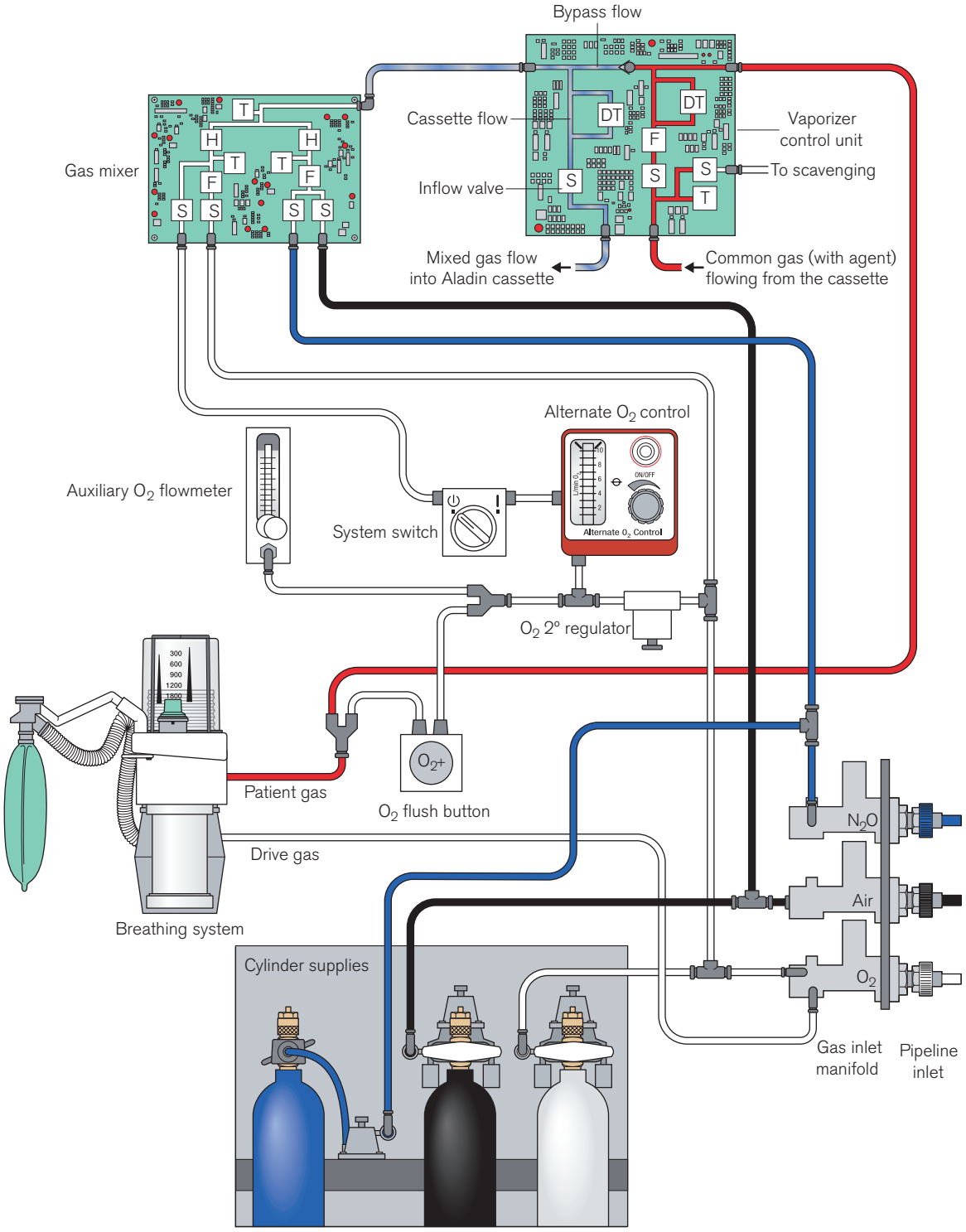


Figure 4.48 Simplified pneumatic system and associated control modules of the Aisys. S, selector valve (solenoid); F, flow control (proportional valve); T, pressure transducer; H, hot wire anemometer (flow meter); DT, differential pressure transducer. The reserve N_2O cylinder is housed separately on the machine and is attached via flexible high pressure tubing to its regulator.
Redrawn with permission of GE Healthcare.



Figure 4.49 Alternate O_2 control flowmeter on the front panel of the Aisys.

is maintained and hence volatile agents can still be administered, though of course there would be no second or carrier gas.

An attraction of electronic gas flow control is its flexibility, demonstrated by an additional feature made available in 2010 that simply requires a software upgrade to the CPU of the workstation and a small change to the respiratory gas module of the patient monitor. End tidal control of oxygen and volatile agent concentration is now possible in this device by creating a feedback loop between the patient gas monitoring module and the vaporizer and gas flow control system. The algorithm controlling this function adjusts the fresh gas-flow composition to rapidly achieve the desired concentration in the exhaled patient gas.

REFERENCES

1. Alexander JP, Watters CH, Dodds WJ, McNeill WE. The Engström Elsa anaesthetic machine. An electronic system for anaesthesia. *Anaesthesia* 1990;**45**:746–50.
2. Medicines and Healthcare products Regulatory Agency. Medical Device Alert, Ref: MDA/2010/052. Anaesthetic vaporizers used to administer volatile agents for the maintenance of anaesthesia – all manufacturers. Department of Health 2010, UK. http://www.mhra.gov.uk/Publications/Safetywarnings/MedicalDeviceAlerts/CON085024
3. Chawla AV, Newton NI. Machine and monitoring failure from electrical overloading. *Anaesthesia* 2002;**57**:1134–5.
4. Schreiber PJ, Schreiber J. Structured alarm systems for the operating room. *J Clin Monit* 1989;**5**:201–4.
5. Westenskow DR, Orr JA, Simon FH, et al. Intelligent alarms reduce anaesthesiologist's response time to critical faults. *Anaesthesiology* 1992;**77**:1074–9.
6. Auditory Alarms in Critical Care Settings: Alarm Bibliography. University of Maryland, Baltimore. http://hfrp.umm.edu/alarms/Alarms_bibliography.htm.
7. Mondor TA, Finley GA. The perceived urgency of auditory warning alarms used in the hospital operating room is inappropriate. *Can J Anaesth* 2003;**50**:221–8.
8. Checking Anaesthetic Equipment 3. The Association of Anaesthetists of Great Britain and Ireland, London 2004. http://www.aagbi.org/guidelines.html.

FURTHER READING

Dain S. Current equipment alarm sounds: friend or foe? [editorial] *Can J Anaesth* 2003;**50**:209–14.

Mondor TA, Finley GA. The perceived urgency of auditory warning alarms used in the hospital operating room is inappropriate. *Can J Anaesth* 2003;**50**:221–8.

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Chapter 5

Breathing systems and their components

Andrew J Davey

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INTRODUCTION

The definition and classification of methods in which inhalational agents are delivered to a patient have undergone a number of changes since the origins of anaesthesia. Most current anaesthetic literature uses the following definitions and classifications.

DEFINITIONS

- *Breathing systems*. In this chapter the term breathing system may be used to describe both the apparatus and the mode of operation by which medical gasses and inhalational agents are delivered to the patient, i.e. a 'Mapleson D type breathing system' (mode of operation) would describe the mode of operation of a 'Bain breathing system' (apparatus). The term *breathing circuit* seems to have reappeared also in this context in current anaesthetic literature. However, strictly speaking, a *circuit* refers to apparatus that allows recirculation of medical gasses and vapours and does not accurately describe other pieces of apparatus used for medical gas delivery
- *Rebreathing*. Rebreathing in anaesthetic systems now conventionally refers to the rebreathing of some or all of the previously exhaled gasses, including carbon dioxide and water vapour. (Rebreathing apparatus in other spheres, e.g. fire fighting and underwater diving, has always referred to the recirculation of expired gas suitably purified and with the oxygen content restored or increased.)
- *Apparatus dead space*. This refers to that volume within the apparatus that may contain exhaled patient gas and which will be rebreathed at the beginning of a subsequent inspiratory breath (Fig. 5.1).
- *Functional dead space*. Some systems may well have a smaller 'functional' dead space owing to the flushing effect of a continuous fresh gas stream at the end of expiration replacing exhaled gas in the apparatus dead space (Fig. 5.2).

CLASSIFICATION OF BREATHING SYSTEMS

These may be classified according to function:

- non-rebreathing systems (utilizing non-rebreathing valves)
- systems where some rebreathing of previously exhaled gas is possible, but normally prevented by the flow of fresh gas through the system

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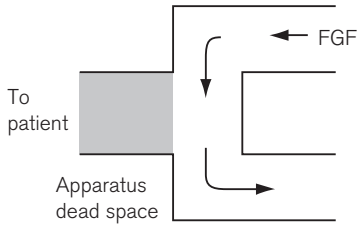


Figure 5.1 Apparatus dead space in a breathing system. FGF, fresh gas flow.

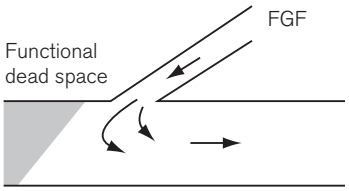


Figure 5.2 Functional dead space in a breathing system with an angled FGF inlet. FGF, fresh gas flow.

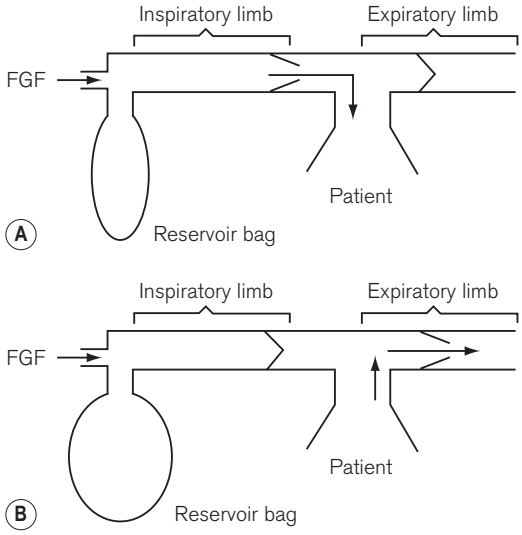


Figure 5.3 Non-rebreathing valve. **A.** Inspiration. **B.** Exhalation. FGF, fresh gas flow.

- non-rebreathing systems utilizing carbon dioxide absorption and recirculation of gasses:
  - unidirectional (circle) systems
  - bi-directional (to-and-fro) systems.

Non-rebreathing systems

The simplest way to deliver a consistent fresh gas supply to a patient is with a system that utilizes a non-rebreathing valve (or valves).

Fresh gas enters the system via an inspiratory limb (Fig 5.3A). This is a length of hosing that has a sufficiently wide bore to minimize any resistance to airflow. It is reinforced so as to prevent collapse from sub-atmospheric pressure either by manufacturing it with corrugations or by bonding a reinforcing spiral onto the outside of the hose. Both of these methods also allow the tube to bend without kinking. The fresh gas entering is either sucked in by the patient's inspiratory effort or blown in during controlled ventilation and enters the non-rebreathing valve. This valve is so constructed that when it opens to admit inspiratory gas, it occludes the expiratory limb of the system (see Fig. 5.3A). When the patient exhales, the reverse occurs, i.e. the valve mechanism moves to occlude the inspiratory limb and opens the expiratory limb to allow expired gasses to escape (Fig. 5.3B).

The inspiratory limb usually includes a bag (1.5–2 litres capacity) that acts as a reservoir for fresh gas. This reservoir contains enough gas to cope with the intermittent high demand that occurs at inspiration. For example, a patient breathing normally (with a minute volume of 7 litres) may well have a tidal volume of 500 ml inhaled over approximately 1 s. This produces an average inspiratory flow rate (not peak flow rate) of 30 l min^{-1}. Without this reservoir in the system, the fresh gas flow rate would have to at least match this figure (probably more, to match the patient's peak inspiratory flow rate) in order to avoid respiratory embarrassment.

The reservoir bag is refilled with fresh gas during the expiratory phase. It can also be compressed manually to provide assisted or controlled respiration since the non-rebreathing valve works equally effectively in this mode as it does for spontaneous respiration.

In the non-rebreathing system described, the fresh gas flow rate *must not be less than* the minute volume required by the patient.

Systems where rebreathing is possible

A miscellany of breathing systems was developed by early pioneers (largely intuitively) that allowed the to-and-fro movement of inspiratory and expiratory gasses within the breathing system. Carbon dioxide elimination was achieved by the flushing action of fresh gas introduced into this breathing system, rather than by separation of the inspiratory and expiratory gas mixtures by a non-rebreathing valve as described above. As it is mainly the flushing effect of fresh gas that eliminates carbon dioxide, these systems retain the potential for rebreathing of carbon dioxide when fresh gas flow rates are reduced.

Mapleson (1954) classified these systems (A to E) according to their efficiency in eliminating carbon dioxide during spontaneous respiration. An F system, the Jackson Rees modification of system E (Ayre's T-piece), was later added to the classification by Willis (1975), (see below).

Mapleson's classification of breathing systems

Fig. 5.4 illustrates a modified Mapleson classification of breathing systems. These all contain similar components but are assembled in different sequences in order to be used more conveniently in specific circumstances. However, as a result, the efficiency of each system is different. They are catalogued in order (A, B, C, D, E, F) of increased requirement of fresh gas flow to prevent rebreathing during spontaneous respiration. System A requires 0.8–1 times the patient's minute ventilation, B and C require 1.5–2 times the patient's minute ventilation and systems D, E and F (all functionally similar) require 2–3 times the patient's minute ventilation to prevent rebreathing during spontaneous respiration.

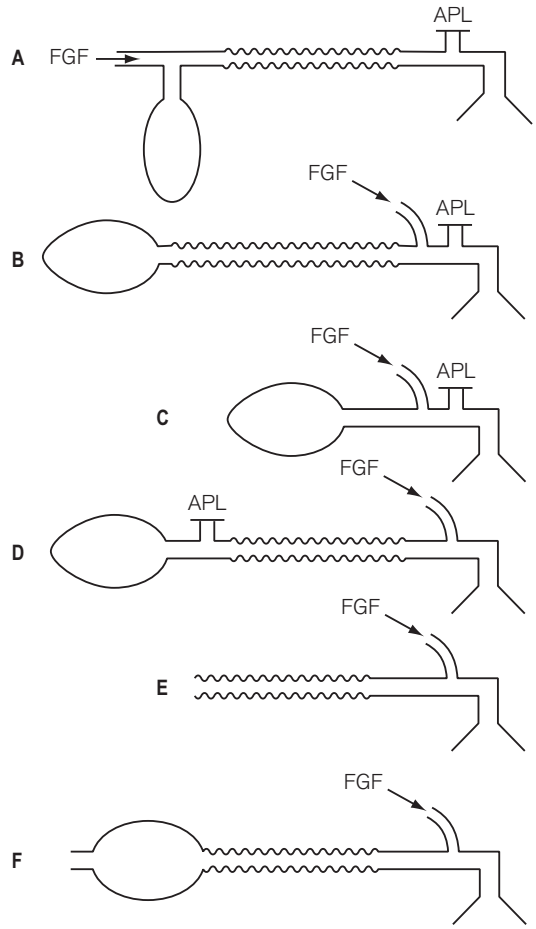


Figure 5.4 A modified Mapleson classification of breathing systems in which there is potential for rebreathing. System **A** houses the gas reservoir in the afferent limb and is alternatively referred to as an afferent reservoir system. The fresh gas flow (FGF) need only be at or just below the patient's minute ventilation without functional rebreathing occurring. Systems **B** and **C** (junctional reservoir systems) require FGF of 1.5 times the minute ventilation to avoid rebreathing. Systems **D, E** and **F** (efferent reservoir systems) require 2–3 times the minute ventilation to avoid rebreathing.

WORKING PRINCIPLES OF BREATHING SYSTEMS

Mapleson A breathing system

The Mapleson A system illustrated in figure 5.5 is the 'Magill attachment' as popularized by Sir Ivan Magill in the 1920s. It consists of the following:

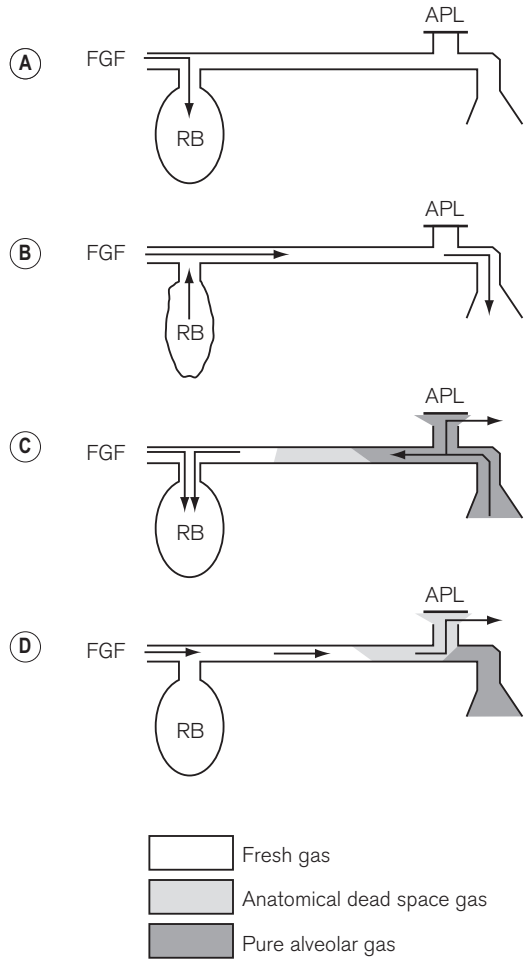


Figure 5.5 The Mapleson A system used with spontaneous breathing (see text). FGF, fresh gas flow; APL, adjustable pressure-limiting valve; RB, reservoir bag.

- At one end, an inlet for fresh gas linked to a 2 litre distensible rubber or neoprene reservoir bag. (Not rebreathing bag, as the patient's exhaled gasses should never be allowed to pass back into it.) This is attached to:
- a length of corrugated breathing hose (minimum length 110 cm with an internal volume of 550 ml). This represents slightly more than the average tidal volume in an anaesthetized adult breathing spontaneously. This volume is important as it minimizes the backtracking of exhaled alveolar gas back to the reservoir bag (see below). This is in turn connected to:
- a variable tension, spring-loaded flap valve for venting of exhaled gasses. This valve should be attached at the opposite end of the system from the reservoir bag and as close to the patient as possible. It will be subsequently referred to as an APL (adjustable pressure limiting) valve.

This system makes efficient use of fresh gas during spontaneous breathing that can be explained by examining its function during a respiratory cycle consisting of three phases: inspiration, expiration, and an end-expiratory pause.

- *The first inspiration.* In Fig. 5.5A the reservoir bag and breathing hose have been filled by an ideal fresh gas flow and attached (with a gas-tight fit) to the patient, who is about to take a breath. The whole system is therefore full of fresh gas. As the patient inspires, the gasses are drawn into the lungs at a rate greater than the fresh gas flow and so the reservoir bag partially empties as shown in Fig. 5.5B.
- *Expiration.* In Fig. 5.5C the patient has begun to exhale, and because the reservoir bag is not full, the exhaled gasses are breathed back along the corrugated hose, pushing the fresh gasses in the hose back towards the reservoir bag. However, before the exhaled gasses can pass as far as the reservoir bag (hence the importance of the length of the inspiratory hose), the latter has been refilled by the fresh gasses from the corrugated hose plus the continuing fresh gas flow from the anaesthetic machine.

  A point is reached when the reservoir bag is again full and, as the patient is still exhaling, the remaining exhaled gasses have to pass out through the APL (expiratory) valve, which now opens.

  The first portion of exhaled gasses to pass along the corrugated hose from the patient was that occupying the patient's anatomical dead space and, therefore, apart from being warmed and slightly humidified (a satisfactory state of affairs), is unaltered, not having taken part in respiratory exchange. This is followed by alveolar gas (with a reduced oxygen content and containing carbon dioxide), the first part of which may enter the corrugated hose, and the rest which is expelled through the APL valve when the reservoir bag is full.
- *End-expiratory pause.* The next stage is the end-expiratory pause. The fresh gas flow entering the system now drives the exhaled gasses, or some that had tracked back along the corrugated hose, out through the APL valve. It can be seen that the expiratory pause is important because it prevents the potential for the rebreathing of exhaled alveolar gasses that would otherwise be contained in the hose at the end of expiration (Fig. 5.5D).

During the end-expiratory pause, all the alveolar gasses and some of the dead-space gasses are expelled from the corrugated hose through the APL valve by the continuing fresh gas flow. Thus, during the next inspiratory phase, the gas inspired may well initially contain some of the remaining dead-space gasses from the previous breath, along with fresh gas. As explained above, these dead-space gasses may be re-inspired without detriment to the patient. The fresh gas flow rate may, therefore, be rather less than the patient's minute volume and rebreathing of alveolar gas is, therefore, prevented.

In theory, provided there is no mixing of fresh gas, dead-space gas and alveolar gas and a sufficient end-expiratory pause, the fresh gas flow rate need only match alveolar ventilation (approximately 66% of the minute volume), as in this situation alveolar gas only will be vented through the APL valve.

In practice, however, a number of factors dictate a higher fresh gas flow rate (70–90%), for instance:

- there is mixing of the various gaseous interfaces, which reduces the theoretical efficiency of the system
- occasionally, larger than expected tidal volumes may well be exhaled and, therefore, reach the reservoir bag, in which case carbon dioxide will contaminate the reservoir bag and the subsequent inspiratory gasses
- rapid respiratory rates will reduce or even eliminate an end-expiratory pause and reduce the potential for carbon dioxide elimination that this pause allows.

Mapleson A system and controlled ventilation

The mechanical aspects of the Mapleson A (Fig. 5.6) system (Magill attachment) as described above relate to its use in spontaneous respiration. However, if controlled or assisted ventilation is used, with the patient's lungs inflated by means of squeezing the reservoir bag, a different state of affairs occurs. With the same fresh gas flows as before, we would see the following:

- *Inspiratory phase.* The APL valve has to be kept almost closed so that sufficient pressure can develop in the system to inflate the lungs. During the first

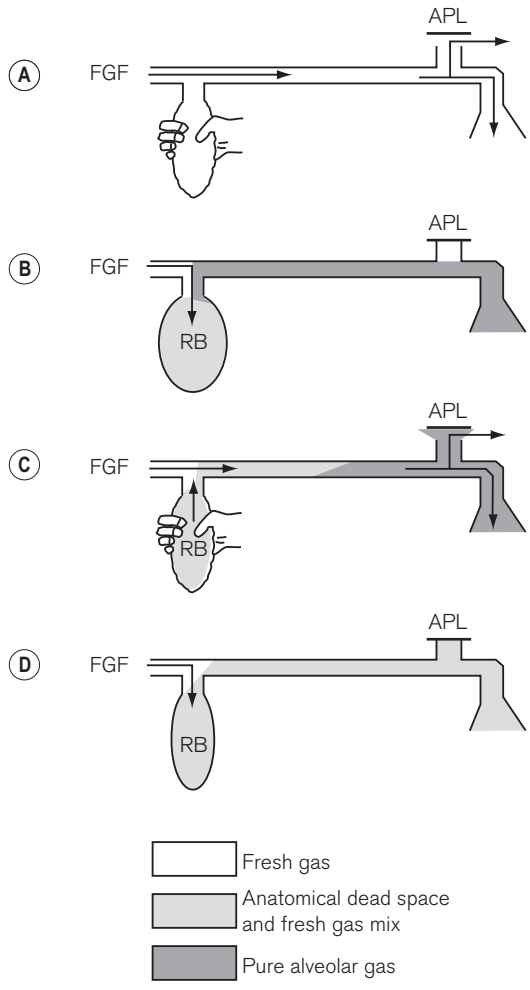


Figure 5.6 Mapleson A system with assisted or controlled ventilation: **A.** at the end of inspiration; **B.** at the end of expiration; **C.** during subsequent inspiration; **D.** at the end of the subsequent inspiration. Note that much rebreathing takes place (see text). FGF, fresh gas flow; APL, adjustable pressure-limiting valve; RB, reservoir bag.

inspiratory phase with the anaesthetist squeezing the bag, some of the fresh gasses are blown out of the valve.

- *Expiratory phase.* At the end of inspiration, the reservoir bag may be almost empty, and as soon as the anaesthetist relaxes his pressure on it, the patient exhales into the corrugated hose. The exhaled dead space and alveolar gasses pass further back along the breathing hose and may even enter the reservoir bag. The bag rarely fills completely and so there is usually insufficient pressure within the system to open the APL valve during this phase.

When the anaesthetist squeezes the bag again, the first gasses to enter the patient's lungs will be the previously exhaled alveolar gasses. The volume of gasses escaping via the APL valve during this second inspiratory phase is initially small (the valve being almost closed), but gradually increases as the pressure in the system rises towards the maximum inspiration. Therefore, the greatest amount of gas will he dumped late in the cycle and will consist mainly of fresh gas. Under these circumstances there is considerable rebreathing (Fig. 5.6 see shading). Furthermore, as alveolar gas will have entered the reservoir bag, there will always be carbon dioxide contamination in any subsequent inspirate. In order to prevent this and thereby minimize the potential for rebreathing alveolar gas, a high fresh gas flow rate is required.

This is usually of the order of 2 times the patient's minute ventilation. This situation is highly wasteful with regard to fresh gas and also increases the potential for pollution.

Other Mapleson A breathing systems

The Lack co-axial breathing system (Fig. 5.7A)

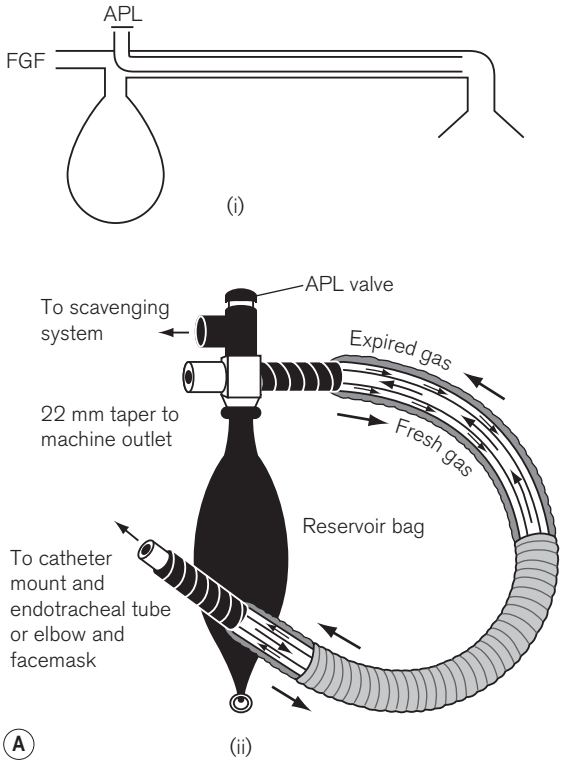
The traditional layout of a Magill system sites the APL valve as close to the patient as possible. However, this:

- reduces the access to the valve in head and neck surgery
- increases the drag on the mask or endotracheal tube when the valve is shrouded and connected to scavenging tubing.

The Lack system overcomes these two problems. The original version was constructed with a co-axial arrangement of breathing hoses. Exhaled gasses are passed into the orifice of the inner hose sited at the patient end of the system and then back towards the APL valve, which is now sited on the reservoir bag mount. The valve is thus conveniently sited for adjustment by the anaesthetist and its weight, and that of any additional scavenging attachment, is now supported by the anaesthetic machine (see Fig. 5.7). The system still functions as a Mapleson A system. The co-axial hosing on early models was criticised as being too narrow and having too high a flow resistance. In later models, the inner and outer breathing hose diameters were subsequently both increased, to 15 and 30 mm respectively, to overcome this problem. Another problem was that the tubing was heavy and stiff, putting a stress on the connection to the facepiece or endotracheal connection.

Lack parallel breathing system (Fig. 5.7B)

Co-axial breathing systems have particular hazards. If the inner hose were to become disconnected or to split, as has been the case, the leak may pass unnoticed. This would



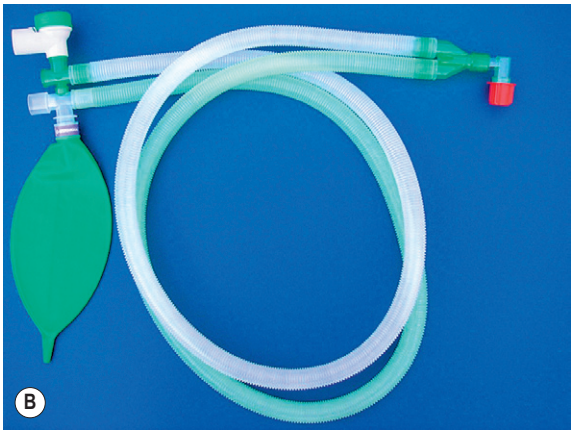


Figure 5.7 **A.** The Lack co-axial breathing system. (i) Working principles. (ii) The actual assembly. The outer corrugated hose is partly transparent so that the inner tube may be seen to be intact. The adjustable pressure-limiting valve (APL) is fitted with a shroud having a 30 mm outlet so that it may be attached to a scavenging system. **B.** The Lack parallel breathing attachment. The inner (expiratory) limb of the co-axial arrangement has been replaced by one that is parallel to the inspiratory limb. FGF, fresh gas flow.

drastically alter the efficiency of the system in eliminating carbon dioxide and is therefore dangerous. A version of the Lack system with parallel hoses is now available (see Fig. 5.7B).

Mapleson B and C systems

These place both the fresh gas supply, reservoir bag and APL valve closer to the patient, allowing the anaesthetist easy access to the latter two, whilst at the same time allowing the fresh gas source to be sited at a distance if necessary. However, as mentioned above, neither is as economical as the 'A' system for spontaneous respiration. The reasons are outlined below for the 'B' system.

- *The first inspiration.* The system is initially assumed to be full of fresh gas so that during the first inspiration the patient breathes only fresh gas.
- *Expiration.* During expiration, the exhaled gasses (initially dead space gas and then the first part of the alveolar gas), mixed with the fresh gas flow (FGF), pass to the reservoir bag. When the latter has been refilled, the remainder of the exhaled gasses (the rest of the alveolar gas) and the FGF are voided via the APL valve.
- *End-expiratory pause.* During this phase, it is fresh gas that escapes from the APL valve as this is closer to the valve than the bag.
- *The next inspiration.* This will be supplied by the contents of the bag which has a mixture of fresh, dead space and alveolar gas, the proportion of which will be determined by the fresh gas flow rate and the rate at which exhalation occurred. If the fresh gas flow is high and the exhalation rate was slow, there will be a greater amount of fresh gas in the inspirate.

Mapleson D system

The Mapleson D system with spontaneous respiration

The system is best explained hypothetically, if again the three phases of the respiratory cycle are considered in sequence: inspiration, expiration and end-expiratory pause.

- *The first inspiration.* An appropriate fresh gas flow (see later) enters as close as possible to the patient end of the breathing system (so as to reduce any apparatus dead space) and the system including the reservoir bag is filled so that during the first inspiration the patient breathes only fresh gas. As the inspiratory rate is greater than the FGF the bag begins to empty (Fig. 5.8A).

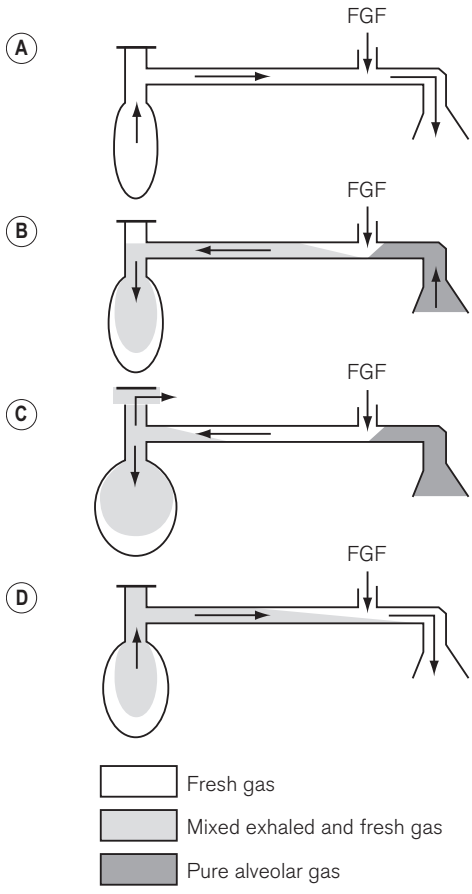


Figure 5.8 The Mapleson D system with spontaneous ventilation (see text). FGF, fresh gas flow; APL, adjustable pressure-limiting valve.

- *Expiration.* During expiration, the exhaled gasses mix with fresh gas entering the system and these pass down the wide bore hose. They initially displace any fresh gas remaining here and start to fill the reservoir bag (Fig. 5.8B). When the bag is full, the remainder of the exhaled gasses and the FGF are voided via the APL (expiratory) valve. Of the expired gasses, it is those from the patient's respiratory dead space that are voided first, followed by alveolar gasses.
- *End-expiratory pause.* During the end-expiratory pause the fresh gas flow entering the system passes down the wide bore hose, displacing some of the mixture of exhaled gas and FGF, which is now vented out through the APL valve (Fig. 5.8C). The amount of fresh gas occupying and thus stored in the patient end of the wide bore hose at the end of expiration, therefore depends on the fresh gas flow rate, the duration of the end-expiratory pause, and the degree of mixing (due to turbulence) of the various gaseous interfaces within the corrugated hose.
- At the next inspiration the inhaled gasses consist initially of this stored fresh gas followed then by the mixture of exhaled gasses and FGF that remain in the tube and possibly some of the mixture from the reservoir bag if the inhaled tidal volume is large (Fig. 5.8D).

However, there are practical problems with this concept, since the expiratory pause may be short, particularly during spontaneous breathing (when volatile anaesthetics only are used with minimal opioid supplementation). In this case the fresh gas flow needs to be sufficiently high to flush the exhaled gasses downstream prior to the next inspiration. In fact, rebreathing of exhaled alveolar gas occurs unless the fresh gas flow is at least 2 times and possibly up to 4 times the patient's minute ventilation.

It is worthy of note that the Mapleson D system is functionally similar to a T-piece (Mapleson E). However, with a T-piece, the limb through which the ventilation occurs, if used without a reservoir bag, must be of such a length that the volume of gas in it when augmented by the volume of the fresh gas flow being delivered during inspiration is no less than that of the patient's tidal volume, otherwise dilution of anaesthetic by entrained air will occur.

Mapleson D system with controlled or assisted ventilation

- *The first inspiration.* As the bag (full of fresh gas) is squeezed, the fresh gas flow entering the system as well as fresh gasses stored in the wide-bore breathing hose pass to the patient. At the same time some gasses from the reservoir bag are lost through the partially open APL (expiratory) valve (Fig. 5.9A).
- *Expiration.* A mixture of the fresh gas flow and exhaled gasses passes along the hose, eventually entering the now partially deflated reservoir bag, causing it to refill (Fig. 5.9B).
- *Expiratory pause.* At this point, provided that there is an expiratory pause, the fresh gas supply continues to flow down the hose to replace and drive the mixed gasses out via the APL valve. A longer expiratory pause allows a greater amount of fresh gas to enter the breathing hose (Fig. 5.9C).
- *The next inspiration.* At the next squeeze of the reservoir bag (Fig. 5.9D) the continuing fresh gas flow, plus the fresh gas now stored in the breathing hose plus any previously expired gasses that may remain in the hose pass to the patient, while some of the mixed gasses within the bag escape via the APL valve. The cycle then repeats itself.

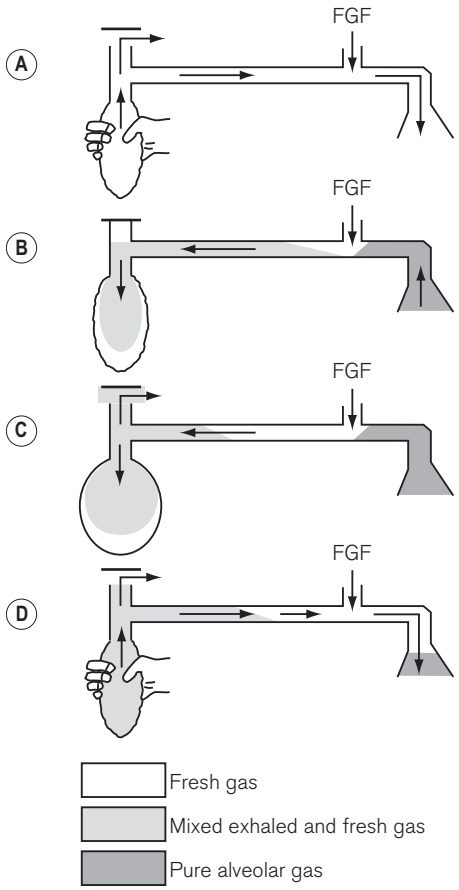


Figure 5.9 Mapleson D system with manual ventilation. **A.** The first inspiration; note that the APL valve is forced open. **B.** Early exhalation; the APL valve is closed and the partially collapsed reservoir bag is filling. **C.** Late exhalation/expiratory pause; mixed gas is vented from the system. **D.** Subsequent inspiration.

Thus, to prevent rebreathing in the Mapleson D system during both spontaneous and controlled ventilation, the fresh gas flow must be sufficiently high enough to:

- purge the breathing hose of exhaled gasses
- supplement the stored fresh gas in this breathing hose so that any mixed gas in the reservoir bag is prevented from reaching the patient. The amount of fresh gas required will always be greater than the patient's minute volume and will depend largely on the expiratory pause. The longer the pause, the more effective will be the ability of the fresh gas to purge the breathing hose of expired gas.

However, in practice during controlled ventilation, *deliberate use is often made of functional rebreathing*. Theoretically, if slow ventilation rates (with long expiratory pauses) and large tidal volumes are chosen, then sufficient expiratory time will elapse to allow a modest fresh gas flow to fill the proximal part of the system with sufficient fresh gas to provide alveolar ventilation. This will enter the lungs first, followed by a mixture of previously expired gasses which will then occupy the patient's anatomical dead space. Hence, theoretically, it should be possible to reduce the fresh gas flow to the volume required for alveolar ventilation.

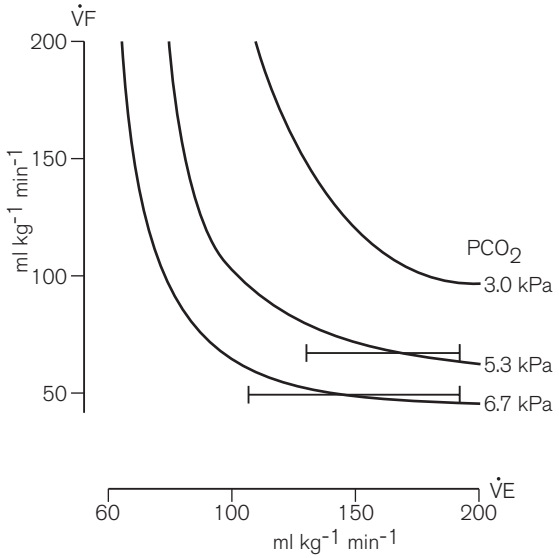


Figure 5.10 Isopleths showing the relationship between fresh gas flow (V_F), minute ventilation (V_E) and various alveolar carbon dioxide tensions.

In practice, there is turbulent mixing of the various gaseous interfaces so that alveolar gas is widely distributed (and diluted). Even so, provided *sufficiently large controlled minute ventilation* is delivered so that most of the fresh gas flow reaches the alveoli, adequate alveolar ventilation will occur with fresh gas flow rates of 70% of the anticipated minute ventilation since, as mentioned above, some rebreathing is acceptable. Fig. 5.10 demonstrates this as well as the fact that the arterial carbon dioxide tension remains fairly constant for any given fresh gas flow rate despite alterations in minute ventilation.

Mapleson D systems are thus able to make efficient use of fresh anaesthetic gasses during controlled ventilation and could have considerable cost saving benefits. Fig. 5.11 shows how the Mapleson D system may be employed with an automatic ventilator. The reservoir bag is removed and replaced with a standard length of corrugated hose of sufficient capacity to accommodate the air or oxygen that is delivered by the ventilator, and therefore prevents it reaching the patient in place of the intended anaesthetic gasses.

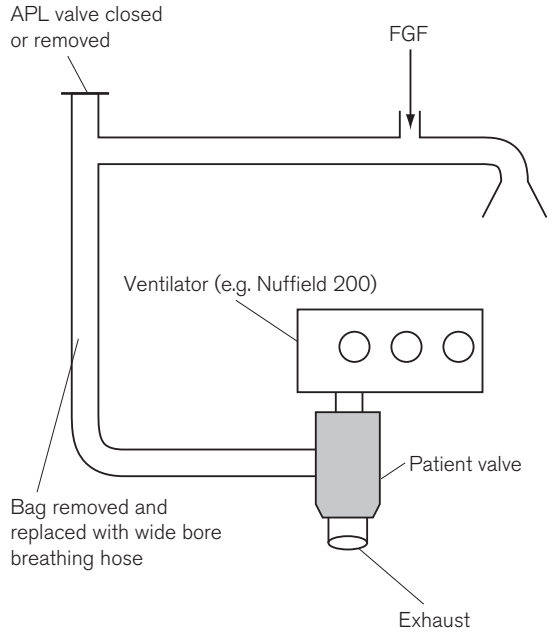


Figure 5.11 Controlled ventilation with a Mapleson D system using a ventilator. Note that with the employment of a ventilator that operates as a 'gas piston', the former must be separated from the breathing system by a suitably long piece of breathing hose (with an internal volume of at least 500 ml). This prevents the driving gas from reaching the patient and diluting the anaesthetic mixture. APL, adjustable pressure-limiting valve; FGF, fresh gas flow.

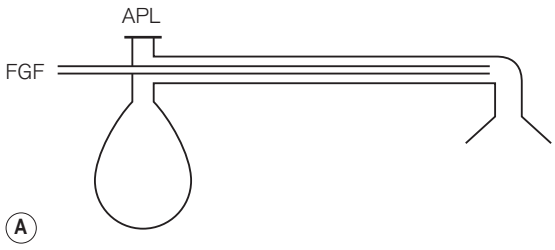
The Bain system

The Bain breathing system (Fig. 5.12) is similar in function to the Mapleson D system. The only difference is that the fresh gas flow is carried by a tube within the corrugated hose (a co-axial arrangement). In the earlier models in particular, there was a risk that the inner tube could become disconnected at the machine end; if this happened a very big dead space was introduced. It could also become kinked, so cutting off the supply of fresh gasses.

Hybrid systems

A number of breathing systems have been described that, by means of a lever switch, can convert the system from a Mapleson A to a Mapleson D or E, allowing a system to be chosen and used in its most efficient mode (i.e. system A, spontaneous respiration; system E, controlled respiration). The Humphrey ADE system (Anaequip) seems to be the most popular version (see Fig. 5.13A). A 'Y' shaped length of corrugated breathing hose is attached to H and F in the diagram with the short limb of the Y at the patient.

With the lever up in the A mode (Fig. 5.13B), the reservoir bag on the ADE block is connected to the inspiratory pathway as in a Mapleson A system. The breathing hose connecting the block at H to the patient is now designated as the inspiratory limb. Expired gas is carried back along the other limb to the block at F and then is vented through an APL valve which is shrouded to facilitate scavenging. In practice it appears to function more efficiently than a traditional Magill attachment. The improved efficiency is thought to relate to the position and design of the



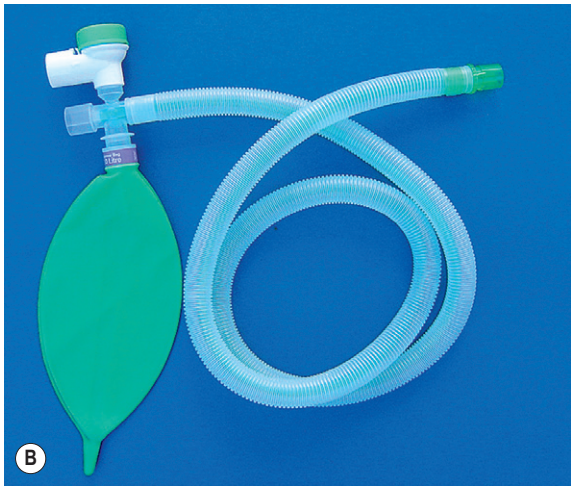


Figure 5.12 The Bain breathing system. **A.** Working principle. **B.** Intersurgical disposable. FGF, fresh gas flow; APL, adjustable pressure-limiting (expiratory) valve.

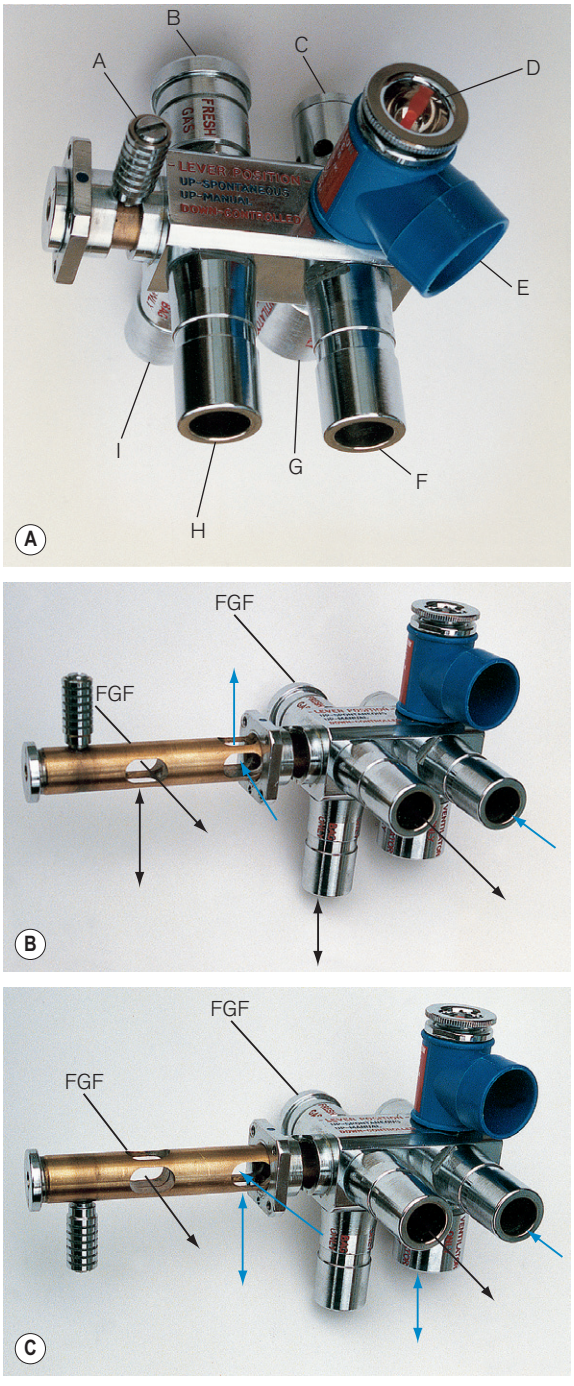


Figure 5.13 **A.** The Humphrey ADE system. A, lever; B, fresh gas input; C, overpressure relief valve; D, APL valve with spindle; E, scavenging shroud; F, 22 mm breathing hose port from patient; G, ventilator port; H, 22 mm breathing hose port to patient; I, bag mount. **B.** An exploded view; with the lever set upright, the system functions in its Mapleson A mode for spontaneous respiration with a recommended fresh gas flow (FGF) of 70 ml kg^{-1} or less if used with capnography. Manual ventilation is easily instituted by pressing on the spindle to close the valve during inspiration and releasing it during exhalation. **C.** For mechanical ventilation the lever is positioned downwards, converting the breathing system to the Mapleson E mode by isolating the reservoir bag and APL valve but incorporating the ventilator port. The fresh gas flow may be kept the same.

components at the patient end of the system. Towards the end of exhalation in a Magill attachment, the exhaled dead-space gas which has passed up the breathing hose is now returned towards the APL valve by the flushing action of the fresh gas flow entering the system. At the APL valve, it meets and mixes with alveolar gas in turbulent fashion, and a mixture of both is discharged from the valve. However, in the Humphrey (and Lack) systems alveolar gas is diverted in a more laminar fashion into a physically separate expiratory limb, which minimizes any potential for mixing of the two gas phases in question. This arrangement, and the removal of the APL valve assembly away from the patient end of the system, also reduces apparatus dead space, so that with further modification (see below) it may be suitable for infants and neonates.

With the lever down in the D/E mode (Fig. 5.13C), the reservoir bag and APL valve are isolated from the breathing system. What was the 'inspiratory' limb in 'A' mode now delivers gas to the patient end of the system as in a T-piece (see below). The breathing hose returning gas to the ADE block now functions as the reservoir limb of a T-piece. This hose vents to atmosphere via a port adjacent to the bag mount. As this mode does not incorporate a reservoir bag, it is strictly a Mapleson E system. The port described above is usually connected to a ventilator of the 'bag-squeezer' type (see Chapter 9), so that the system can be used in its most efficient mode for controlled ventilation.

The Anaequip version is supplied with 15 mm smooth bore non-kinking breathing hose and a unique APL valve (see later in chapter). Interestingly, the use of this smooth bore hose reduces turbulence in the range of flows seen in quietly breathing adults, so that its performance is little different from that of 22 mm corrugated hose. The narrower bore hose also reduces the internal volume of the system to an extent that it is now also suitable for use with infants. A low internal volume is important in a paediatric breathing system in order that:

order that these bases and carbon dioxide (as carbonic acid) can exist in ionic form.

Chemical composition of absorbents

The main constituent of absorbents made by different manufacturers is calcium hydroxide. Other constituents may be added to enhance the reactivity. These include the following:

Sodium/potassium

Small quantities of sodium hydroxide (1.5–5%) can be added to enhance the reactivity and the hygroscopic property (ability to bind water) of the mixture, hence the reason that absorbents containing this are often referred to as 'soda lime'. Some manufacturers have added potassium hydroxide for similar reasons, although this has been discontinued (see below).

Barium

Barium lime conventionally is made up of 85% calcium hydroxide, 11% barium hydroxide and 4% potassium hydroxide. The barium hydroxide has its own naturally occurring 'water of crystallization' and in combination with calcium hydroxide has never required additional hardeners (see later). However, products containing barium were removed from the market in autumn 2004.

Water content

The optimal moisture content of the absorbent mixture is between 14 and 16%. This is essential for the classic chain reaction set out below.

$$\underset{\text{Carbon dioxide}}{CO_2} + \underset{\text{Water}}{H_2O} = H^+ + \underset{\text{Carbonic acid}}{HCO_3^-}$$

$$\underset{\text{Sodium hydroxide}}{2NaOH} + H^+ + \underset{\text{Carbonic acid}}{HCO_3} = \underset{\text{Sodium carbonate}}{Na_2CO_3} + \underset{\text{Water}}{2H_2O}$$

$$\underset{\text{Sodium carbonate}}{Na_2CO_3} + \underset{\text{Calcium hydroxide}}{Ca(OH)_2} = \underset{\text{Calcium carbonate}}{CaCO_3} + \underset{\text{Sodium hydroxide}}{2NaOH}$$

In equation 2, sodium may be substituted by potassium when considering the reaction of absorbents containing the latter.

The reaction is interesting in that:

- it produces heat energy (it is an exothermic reaction)
- it changes the pH of the soda lime, which allows the use of indicator dyes to show when the absorbent is exhausted
- it produces more water than that used up in the reaction. In fact, for each mole of carbon dioxide absorbed one mole of water is produced.

There are other unwanted reactions that take place (associated with dry soda/barium lime); these are discussed further on in the chapter.

Other constituents

Zeolite

Zeolites are three-dimensional, microporous crystalline solids that contain aluminium, silica and oxygen in their regular framework; cations and water are located in the pores. Zeolites have void space (cavities or channels) that can host other molecules. They are naturally occurring minerals that are mined in many parts of the world; however, most commercial zeolites are produced synthetically. They may be added to an absorbent (Spherasorb) to increase:

- *Porosity of the mixture*. The absorbent behaves more efficiently as it becomes more porous due to the increase in available surface area
- *Hardness*. The harder the mixture the less likely it is to form dust. This may be carried around the breathing system and is caustic if inhaled. It may also become damp and settle on valves within the breathing system and cause them to become sticky
- *Water content*. Zeolite also helps to prevent the drying out of the absorbent in adverse conditions. However, should this occur, the zeolite may start to absorb volatile anaesthetic thereby reducing the concentration in the breathing system especially at the beginning of an anaesthetic. Up to 80% of the agent may be absorbed in this situation.

Silica

Sodium and potassium hydroxides have traditionally been added to increase the reactivity of absorbents and to provide the hygroscopic (water retaining) properties. However, they are the main cause of degradation by dry soda lime of anaesthetic agents (isoflurane/desflurane) to carbon monoxide and the degradation of sevoflurane to compound A, formaldehyde and methanol. Calcium hydroxide was thought to be insufficiently reactive on its own to be a suitable agent. However, new methods of production have led to significantly more porous granules (see above) so that calcium hydroxide alone may now be sufficient as an absorbent. However, the granules are much softer as a result and require the addition of silica (LoFloSorb) to overcome this. The new absorbents have been shown to minimize the production of the unwanted compounds above.

Calcium chloride

The addition of 3% calcium chloride (AmsorbPlus) in place of sodium or potassium has a similar, but less powerful effect. This product also contains 1% calcium sulphate to improve the hardness of the granule. Unlike absorbents that contain strong bases, there may also be a colour change (similar to the exhausted state) should a sample dry out. Should this occur, the mixture still prevents the formation of carbon monoxide (with isoflurane and desflurane) and compound A (with sevoflurane).

Granule size

The size of the granules is important. Too large a granule size produces large gaps in a canister of stacked granules, leading to poor contact with gasses passing through and consequent inefficient absorption of carbon dioxide. Too small a granule may provide an unacceptably high resistance to gas flow, along with the increased possibility of dust formation. The optimum size of granule is thought to be between 1.5 and 5 mm in diameter.

Production

The ingredients are mixed along with added water into a paste. This is treated in a number of ways. Originally the paste was dried and then crushed between rollers so that it formed granules. The product was then sieved through various meshes to retain the sizes quoted above. The dried granules were then sprayed with water to give the right amount of water content to allow optimal reactivity without making the granule soft or sticky. Mesh standards differ between countries due to variations in thickness of the wire used to construct the mesh. In the USA soda lime granules are supplied at between 4 and 8 Mesh USP (2.36–4.75 mm). In the UK the granules are supplied to a BP (British Pharmacopoeia) standard of 1.4–4.75 mm (3–10 Mesh). More recently, production methods have been introduced that produce a more uniform size. The paste may be squeezed through a sieve like spaghetti and chopped into little pieces. It may be passed through a mangle that has thousands of dimples on its surface. This produces tiny similar-sized spheres that are blown off the roller by a high-pressure jet of air. The paste may also be placed on a dimpled belt that is passed through a smooth roller. This produces little hemispheres that are removed from the belt in a manner similar to removing an ice cube from its tray.

The dust is caustic and can produce burns in the respiratory tract if inhaled. This was a problem with the older type of system with 'to-and-fro' absorption (Waters' Canister, see below), where the absorber was placed in close proximity to the patient's airway. However, circle absorbers are usually separated from the patient by at least 1 m of breathing hose, which normally hangs in a loop. This allows the dust, if present, to fall out before it call get to the patient. Furthermore, a breathing filter separating the patient from the breathing system would protect against this.

Absorptive capacity

Traditional soda lime is capable of absorbing 25 L of carbon dioxide per 100 g, and barium lime 27 L of carbon dioxide per 100 g. However, in continuous use, the absorbent appears exhausted (as indicated by the colour change) before these capacities are reached, because the outside of the granule is exhausted before the whole granule is used up. Furthermore, the contact time between the absorbent and carbon dioxide affects efficiency. Smaller canisters containing 500 g of absorbent appear exhausted at a carbon dioxide load of 10–12 L per 100 g. 'Jumbo' absorbers containing 2 kg of absorbent, which allow a longer contact time between it and the carbon dioxide, appear exhausted at a carbon dioxide load of 17 L per 100 g.

When the system is allowed to stand for a few hours, absorbent appears to 'regenerate' as the surface carbonate is diluted by hydroxide ions migrating from within the granule. The colour of fresh absorbent depends on which indicator dye is added by the manufacturer. For example if 'Titan yellow' is added, the product turns from deep pink when fresh to off-white when exhausted, whereas if 'ethyl violet' is used as the indicator dye, the colour changes from white when fresh to purple when exhausted. It is, therefore, important to know which colour change to expect. Some absorbents (LoFloSorb) have a green pigment added to the ethyl violet so that it is pale green when fresh, but violet when exhausted.

The exothermic reaction

The heat and water produced by the reaction of absorbent on carbon dioxide has been considered to be beneficial in that (at low flows) they warm and partially humidify the inspiratory gas. The temperature and humidity of the inspired gas is related to a number of factors:

- If the fresh gas flow rate is high, the dry gas entering the system reduces both the humidity and the temperature of the recirculating gas.
- At low fresh gas flows, if the gas circulation time is high, the humidity and temperature rises.
- The longer the system is in use at low fresh gas flows, the greater the humidity and temperature of the circulating gas.

The heat produced, however, is not necessarily all beneficial. The chemical reaction between volatile anaesthetic agents and absorbents containing sodium or potassium increases in proportion to the temperature within the system:

- Trichloroethylene can be decomposed to dichloroacetylene (which is neurotoxic) and further to phosgene, if the temperature within the product exceeds 60°C. Older anaesthetic machines often incorporated a switching system on the back bar that could divert gas directly into a circle system rather than to the common gas outlet. The trichloroethylene vaporizer (where fitted) was always positioned downstream of this switch so that it could never be used with the circle absorber system.
- Anaesthetic agents with the CHF_2 moiety (desflurane, enflurane and isoflurane) react with dry, warm soda lime or barium lime (Baralyme) to produce varying amounts of carbon monoxide (Table 5.1).

Table 5.1 Carbon monoxide production by reaction of anaesthetic agents that have the CHF_2 moiety (desflurane, enflurane, isoflurane) with lime

TYPE OF LIME		CO (ppm)
Dry soda lime (<2% water)	Pottasium hydroxide	50-100
Dry soda lime (<2% water)	Sodium hydroxide	15
Fresh soda lime	Sodium hydroxide	<8

Dry barium lime, which contains potassium hydroxide, has a much greater tendency to produce carbon monoxide than dry soda lime that contains sodium hydroxide. Fresh absorbent, which has an approximately 15% water content, appears to prevent carbon monoxide formation. In fact significant production takes place only when the water content drops below 2%. The problem can occur when the absorber is left unused for long periods, or when large amounts of dry gas are allowed to pass through it between cases, particularly overnight or at weekends. Even with the flow meters turned off, this is seen albeit after a long period of time in some anaesthetic machines, if:

- They are plugged into the pipeline supply
- They have a minimal basal flow of oxygen through the machine
- The absorber is so constructed that fresh gas flows through the absorbent prior to entering the inspiratory limb.

Carbon monoxide is not easily measured as an exhaled gas as it binds to haemoglobin in the blood first. Its production in a system using absorbents may be difficult to detect. There is not necessarily a colour change in absorbents containing strong bases (potassium/sodium) if they dry out! Suspicion should be raised if there appears to be excessive heat production from the absorbent. This phenomenon can easily be avoided by:

- using smaller absorbers (so that the contents have to be changed regularly)
- disconnecting them when possible
- only having breathing system designs that allow the fresh gas inflow to bypass the absorbent, and unplugging machines from the pipeline supply when not in use for extended periods.

Sevoflurane may undergo degradation within the absorber, to non-toxic fluorinated metabolites in humans (mainly a sevo-olefin called 'compound A'). Levels rise, as would be expected, with increased concentration of the agent, prolonged anaesthesia, low fresh gas flows and increased operating temperature within the absorber. However, there is no evidence of any danger to humans. In the presence of dried absorbent (especially barium lime) there may be an extreme temperature rise and a number of other breakdown products are produced including formaldehyde and methanol. There have been isolated reports of fire or extreme heat in circle systems resulting in at least one fatality. This prompted the manufacturer to issue a letter to healthcare professionals in the USA warning of the danger. In the cases investigated the following signs were noted:

- failed induction or inadequate anaesthetic depth
- clinical sign of airway irritation
- oxygen desaturation, increased airway pressures
- severe airway oedema and erythema
- elevated carboxyhaemoglobin levels.

In summary the recommendations for preventing the problem are as follows:

- If there is any suspicion that the absorbent has not been in use for an extended period of time it should be replaced.
- The machine should be turned off when not in use so that there is no flow of gas through the absorber.
- The vaporizers should be turned off when not in use.
- The sevoflurane concentrations dialled up on the vaporizer and that measured by an analyzer should be compared and any major disparity should be investigated.
- A check for excessive heat production from the absorbent should be performed periodically.

It is worth noting that as a result of increased awareness and the availability of safer absorbents, there have been no reported incidents of any of the problems discussed above since the year 2000.

CLASSIFICATION OF BREATHING SYSTEMS UTILIZING CARBON DIOXIDE ABSORPTION

Carbon dioxide absorption can be used in two types of system:

- a 'to-and-fro' absorption system
- a circle absorption system.

'To-and-fro' absorption systems

The Waters' canister (Fig. 5.16)

Here the patient breathes in and out of a closed bag, which is connected to the facemask or endotracheal tube via a canister containing soda lime. The part of the system between the patient and the soda lime is dead space and, therefore, its volume must be kept to a minimum. This means that the soda lime canister must be close to the patient's head, and this leads to practical difficulties. A

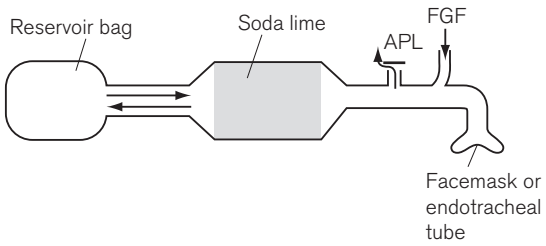


Figure 5.16 A 'to-and-fro' system incorporating a Waters' canister. FGF, fresh gas flow; APL, adjustable pressure-limiting valve.

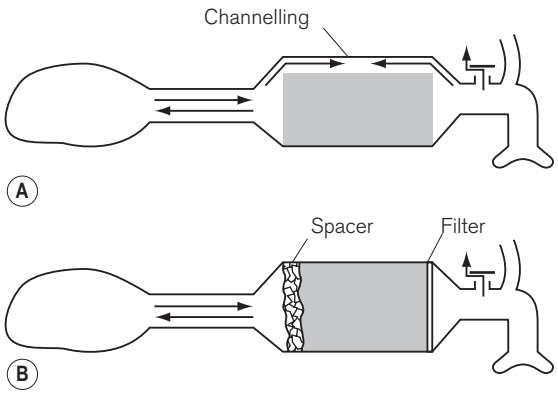


Figure 5.17 **A.** Channelling in a Waters' canister. If the canister is not completely filled with soda lime and is placed in a horizontal position, the gasses can pass through the void at the top and therefore fail to come into adequate contact with the soda lime. **B.** The prevention of channelling by the insertion of a spacer to compress the soda lime. Note also the filter at the patient end that prevents particles of soda lime reaching the patient.

length of wide-bore tubing may, however, be interposed between the canister and reservoir bag without detriment. The fresh gasses are introduced at the patient end of the system, and the expiratory valve is usually mounted close by, though it may equally well be put at the bag end. The canister is usually placed in the horizontal position for convenience, and it is most important that it is well packed, since if there were a space above the soda lime, 'channelling' would occur and absorption would be incomplete (Fig. 5.17A).

Furthermore, the absorbent at the patient end of the system becomes exhausted first and so increases the functional dead space of the system.

Canisters are available as pre-packed, disposable units. In those intended for reuse, the absorbent may conveniently be compressed to prevent gaps by the insertion of a spongy 'spacer' at one end (Fig 5.17B). When the canister is closed, the sealing washer should be checked to ensure that it is in the correct position and any soda lime on the threads of the canister or the sealing washer should be carefully removed as these may cause leaks. The whole system should be tested before use.

Apart from being cumbersome, the 'to-and-fro' system has the disadvantage that the patient could inhale absorbent dust. A breathing filter should be inserted in the patient end of the canister to prevent this.

Circle absorption systems

Here the disadvantages of the absorbent canister being so close to the patient are avoided. The patient is connected to the absorber by two corrugated hoses, one inspiratory and the other expiratory, as shown in Fig. 5.18A. The one way or 'circle' flow of gasses through the system is determined by two unidirectional valves V_1, and V_2, which are accommodated in transparent domes so that their correct action may be observed.

The fresh gas port and the reservoir bag may be sited in the inspiratory pathway close to the inspiratory valve V_1 or in the expiratory limb of the system downstream of the valve V_2. There are claimed advantages for each arrangement. Positioning the FGF and bag in the inspiratory pathway may reduce the resistance to inspiratory effort and prevents desiccation of the absorbent if the fresh gas flow is left on for long periods when the system is not in use. This has implications for the efficiency of the absorbent and its potential for producing unusual breakdown products (see above). However, positioning the FGF and bag in the expiratory limb so that the FGF passes through the absorbent before reaching the patient allegedly warms and humidifies this gas. This also seems to remove some of the excess moisture often seen in absorbers that might increase resistance through the system. The APL valve is usually mounted downstream of the valve V_2 (position 'B') in the expiratory limb, but before gas entry into the absorber. Here, it can dump excess exhaled gas prior to entry of gas into the absorber. At one time it was considered that an APL valve was best positioned in the breathing system at position 'A' for use with spontaneous respiration (Fig. 5.18A). This would preferentially dump alveolar gas during exhalation, thus increasing carbon dioxide elimination upstream of the absorber and conserving soda lime. However, as scavenging assumed a greater importance, the inconvenience of connecting a cumbersome scavenging hose to a valve in this position has limited its usefulness.

Circle breathing systems are manufactured in many different designs and sizes. Most are two part systems with one part comprising a fixed body containing the gas pathways, switches, valves and the other a detachable canister that contains the absorbent. Figs 5.18B, C and D show commercial versions of the system described. Fig. 5.18B (circle system for ADU anaesthetic machine, Datex-Ohmeda) has a container for 1 kg of absorbent that may be either disposable or refilled. During replacement, self-sealing valves on the body of the unit close to prevent

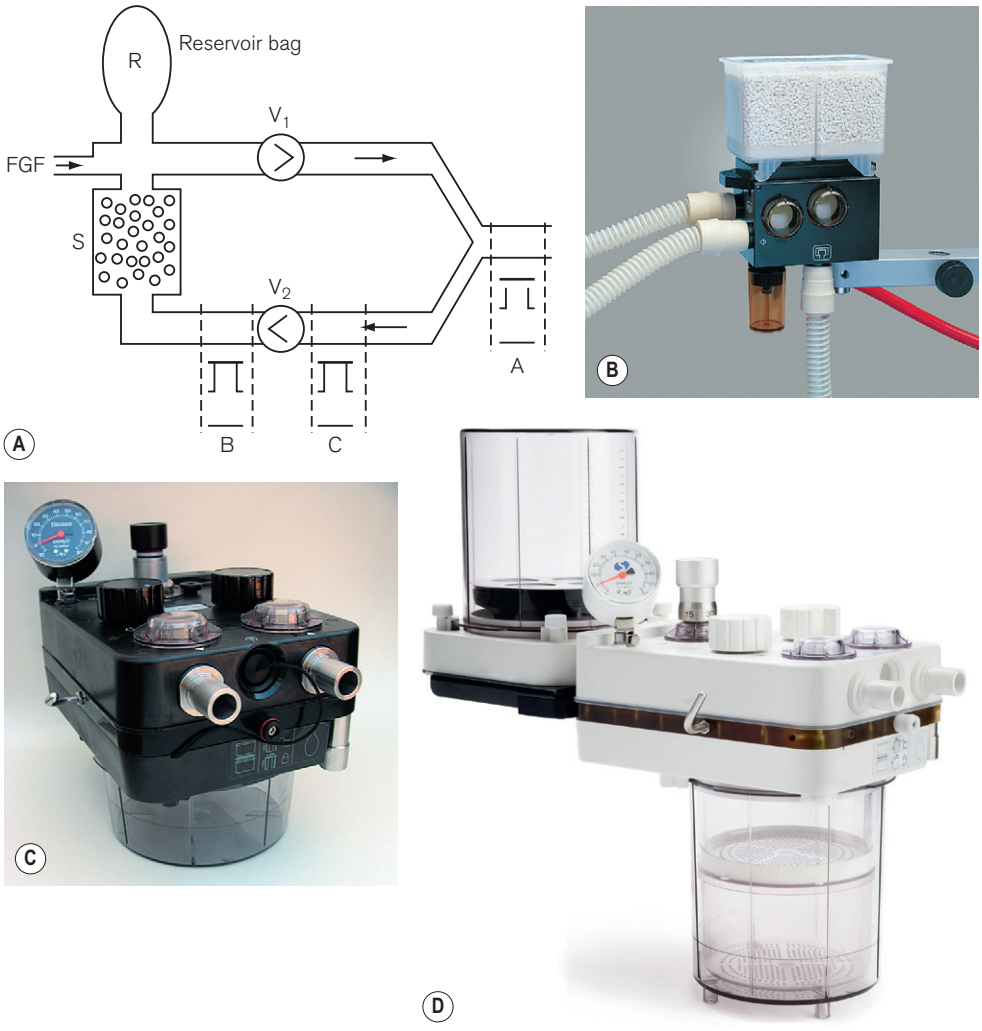


Figure 5.18 **A.** Schematic diagram of a circle breathing system with absorber. FGF, fresh gas flow; V_1 and V_2, one-way valves; R, reservoir bag; S, soda-lime canister. The diagram highlights the alternative siting of the adjustable pressure-limiting (APL) valve at points A, B or C. **B.** ADU circle system, Datex-Ohmeda. **C.** Blease 1 kg absorber. **D.** 2 kg absorber assembly with attached ventilator bellows (Spacelabs).

escape of patient gas from the rest of the system. Figs 5.18C and D show other types of 1 kg and 2 kg absorber respectively. The rationale for larger absorbers is discussed below.

The system shown in Figs 5.19A and B is a disposable version. The valve V_1 is sited in the breathing hose and as close to the patient as possible. In this position it is alleged to have a faster response to pressure changes caused by exhalation and closes earlier, although, as it is exhaled dead space gas that enters the system first, it is immaterial as to which limb this enters initially. Unlike many circle systems that are now an integral part of the anaesthetic workstation, the APL valve is not automatically excluded from the system in mechanical ventilation mode. Therefore, when a ventilator is attached to the reservoir bag port and is in use, the APL valve must be closed fully or gas will be lost from the system.

Apparatus dead space

The apparatus dead space is low in circle systems. It consists only of that volume inside the male taper at the end of the Y piece that joins the inspiratory and expiratory breathing hoses to the patient. However, the functional dead space of this system may vary if a fault develops in it. For example, if the unidirectional valves malfunction and do not fully close, rebreathing can occur from the expiratory limb.

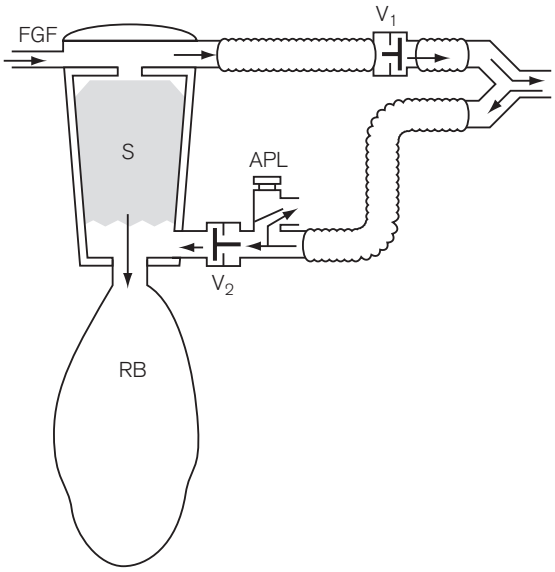


Figure 5.19 The Intersurgical disposable circle system. FGF, fresh gas flow; V_1 and V_2, one-way valves; APL, adjustable pressure-limiting valve; RB, reservoir bag; S, soda lime canister.

Flow resistance

Circle systems impose a greater resistance to breathing than other commonly used breathing systems (Mapleson A-F systems), although less than co-axial arrangements of D systems (Bain system). This is due largely to the fact that there are two extra valves and absorbent in the system (see above). Other factors affecting resistance include the following:

- A high fresh gas flow will assist flow in the inspiratory side of the system, thus decreasing any inspiratory resistance, but increasing expiratory resistance through the unidirectional and APL (expiratory) valves.
- The reverse occurs with low gas flows. Low fresh gas flow rates will also increase the relative humidity and thus increase the 'stickiness' of the unidirectional valves owing to water vapour condensation; therefore further increasing flow resistance.
- The flow rates developed by rapid respiratory excursion (tidal volume and rate) produce the greatest swings in flow resistance (see Table 5.2). These factors may not matter in healthy adults, but they can be unacceptable in young children.

Table 5.2 Flow resistance in circle breathing systems

FREQUENCY (min^{-1})	TIDAL VOLUME (ml)	PRESSURE SWING (cm H_2O)
12	500	–1+½
12	1000	–1 + 1
12	1600	–2+1½
24	500	–1 + 1
24	1000	–3+1½
44	500	–4 + 3

Reproduced from Young TM. Carbon dioxide absorber *Anaesthesia* 1971;**26**:78. With permission.

Efficiency of absorbers

The efficiency of carbon dioxide absorption in a canister depends on:

- the freshness, composition and the available surface area of the absorbent (see above)
- the length of time the gas to be treated is in contact with the granules.

Early canister designs contained approximately 480 g of absorbent. These required frequent changes (after approximately 2–2.5 h of continuous use at low fresh gas flows). Many presently used absorbers are of the 'Jumbo' type, which contain 2 kg of soda lime and, since this has a large volume and surface area of granules, the expired gas is in contact with them for a relatively long period of time, so increasing the efficiency of absorption. It has been shown that a 2-kg canister lasts five times longer than a 0.5-kg canister. When a 2-kg canister is employed it usually has two chambers, one above the other. When one half appears exhausted it is refilled and the canister positions reversed so that the previously unused half now bears the brunt of absorption. Not only is the absorption more efficient in the larger absorbers, but also less frequent recharging is necessary.

With the recent introduction of routine expired carbon dioxide monitoring, these last two considerations appear to be less of a problem in clinical practice, and the reintroduction of smaller absorbers may well have advantages. These are easier to maintain, use and keep clean, and they have fewer leaks and are less prone to causing degradation of volatile agents (see above). The absorbent can also be supplied in disposable cartridges (see Fig. 5.18B).

Absorber switch

Traditionally, an absorber bypass switch has been included in a circle system. This allows expiratory gas to be channelled either through the absorption chamber or across a bypass directly into the inspiratory limb.

Some newer European and American models omit this switch. The rationale for excluding this switch involves safety considerations. Omitting the switch erases the possibility of its inadvertently being left in the absorber

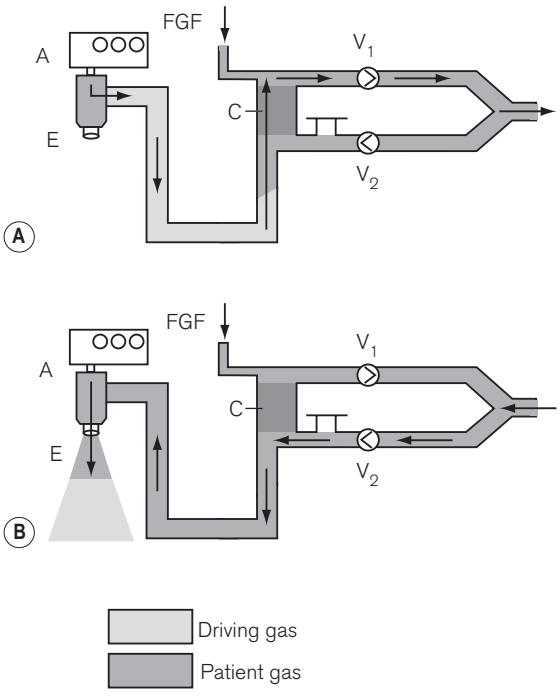


Figure 5.21 Pneumatic piston ventilator and circle system. **A.** Inspiratory phase; **B.** expiratory phase. A, ventilator; C, absorber; V_1 and V_2, one-way valves; E, exhaust; FGF, fresh gas flow.

intended is entered on the ventilator control panel and is delivered to the patient. Transducers monitor the flow to the patient and make adjustments for changes in compliance, resistance and alterations in fresh gas flow so that the tidal volume remains the same.

In older designs that are now mostly obsolete there is no such integration. Here, although the ventilator may be set to give a known tidal volume, when this reaches the circle it is supplemented by the fresh gas entering the system, which may significantly augment the final tidal volume. An example is outlined below:

Consider the following for a FGF entering the circle system:

$$\text{FGF } 6\ \text{l min}^{-1} = 6000\ \text{ml}/60\ \text{s}$$
$$= 100\ \text{ml/s}$$

If ventilator rate $10\ \text{min}^{-1} = 1$ cycle every 6 s
If the insp./exp. ratio is $1/2 = 2$ s insp./4 s exp.
Then insp. FGF in 2 s = **200** ml

If ventilator setting = tidal vol. 400 ml
Total tidal vol. = vent tidal vol. + insp.FGF
= **400** + **200**
= **600** ml

As the FGF is reduced, its contribution to the total tidal volume (and therefore minute ventilation) is reduced and end tidal carbon dioxide is increased in tandem.

Maintenance of circle absorber systems

Prior to the routine use of breathing filters and during prolonged use of the absorber at low fresh gas flow rates, water vapour from exhaled gasses and from the absorbent would condense in the expiratory hose and this needed to be emptied from time to time. Condensation also settled on the unidirectional valves which occasionally caused the valve disc to either adhere by surface tension to the seating or to the cage holding it in place. The former increased flow resistance and the latter allowed rebreathing of CO_2. Nowadays, exhaled water vapour does not usually pass back through the breathing filter into the circle system and this does not seem to pose the same problem.

In parts of the world where the cost of single-use breathing filters would be prohibitive, the tubing should be autoclaved if suitable or washed and hung out to dry between cases. Secondly, the expiratory valve should be dismantled and wiped clean with isopropyl alcohol. When, after dismantling, the glass dome of the expiratory valve is screwed back on again, it is important to ensure that the sealing washer is correctly in place, otherwise a serious leak may occur. If a low-resistance bacterial filter is not used, then the circle absorber housing should be autoclaved (where possible) on a regular basis. Some circle absorbers cannot be autoclaved, but may be cleaned by chemical means.

The absorbent should be changed at regular intervals either when:

- the dye indicates that the majority of the granules are exhausted, or
- when using an analyser, carbon dioxide appears in the inspiratory mixture, or
- when the absorber is unlikely to be used for some time (e.g. over a weekend).

The container for the absorbent usually has a mark above which it should not be filled. Overfilling may result in granules of absorbent clogging the threads of canisters that screw into position, or may prevent the correct seating of the sealing washer, thus causing a leak or bypassing of the absorbent. Furthermore, leaving this space at the top reduces the preferential 'channelling' effect of the gas stream along the sides, and ensures a more even flow through the container. Since the canister is held in the vertical position, channelling is less of a problem than in the Waters' canister, although some does occur between the granules and sides of the canister as the air spaces here are bigger than those between granules within the canister.

Gas and vapour concentration in a circle system

Circle systems are unique in that they function effectively (when a steady state of anaesthesia has been reached) using a wide variety of fresh gas flow rates. However, the fate of the various gasses within the system needs to be understood in order that it may be used safely and effectively. For example, the internal volume of the apparatus (when using a 2 kg absorber, which consists of the intergranular air space in the absorber (1 L), the breathing hoses (1 L) and pathways within the absorber (1 L), totalling 3 L, along with the functional residual capacity of a patient of 1.25 L, provides a large reservoir into which the anaesthetic gas is diluted at the beginning of anaesthesia.

In order to minimize this dilution and provide adequate concentrations of anaesthetic agent, high flows of fresh gas and vapour are required initially in order to flush the residual gas out of the circle system; the higher the flow, the faster this 'wash out' occurs. Lung 'wash out' will of course depend on the patient's minute ventilation. The greater this is, the less time the process takes.

Secondly, the alveolar uptake of anaesthetic agent is greatest at the beginning of anaesthesia. Therefore, the higher the initial fresh gas flow rate (up to a value equal to the patient's minute ventilation) the greater is the delivery of anaesthetic agent into the system. This in turn minimizes any reduction in concentration of agent caused by uptake by the patient. When a near equilibrium of anaesthetic agents in the alveoli and the blood has been reached, exhaled agent concentration almost equals that in the inspiratory mixture, and, therefore, the high initial fresh gas flows may be greatly reduced safely.

In practice the fresh gas flow is usually reduced in stages.

First stage

The initial flow for patient and breathing system washout, as well as supply of adequate anaesthetic agent to match alveolar up-take, usually takes approximately 5–10 minutes. (The shorter time is suited to the insoluble anaesthetic agents desflurane and sevoflurane that reach alveolar equilibrium more quickly.) If this flow rate is at or greater than the patient's minute ventilation, then most or all of the exhaled gas will leave the system via the APL (expiratory) valve without passing through the absorber. When used in this mode it may often be referred to as a *high flow system*.

Second stage

An intermediate flow rate of 70% of the minute ventilation for a further 5 min will allow purging of the soda lime canister without major changes in anaesthetic concentrations.

Third stage

A lower flow may be selected, the value of which will depend on the availability of gas and vapour monitoring within the system, the efficiency of the vaporizer (in or out of circle) and personal preference. Flows of the order of 0.5–2 l min^{-1} are commonly used and when the circle is used in this mode it may often be referred to as a *low flow system*. A *closed flow system* is defined as that which has no gas exit (i.e. APL valve fully closed) and in which the fresh gas flow equals the uptake by the patient. In practice, however, few commercial anaesthetic circle breathing systems are sufficiently leak proof to be used at such flows.

Oxygen concentrations in circle systems at low fresh gas flows

As the fresh gas flow in a circle is decreased, exhaled gas that is allowed to recirculate exerts an increasing influence on the subsequent inspired gas mixture. The oxygen concentration of this exhaled gas depends on:

- its original inspired concentration
- the alveolar oxygen extraction, which may be unpredictable.

Fig. 5.22 shows the decrease in alveolar oxygen concentrations of a 50% nitrous oxide and 50% oxygen mixture

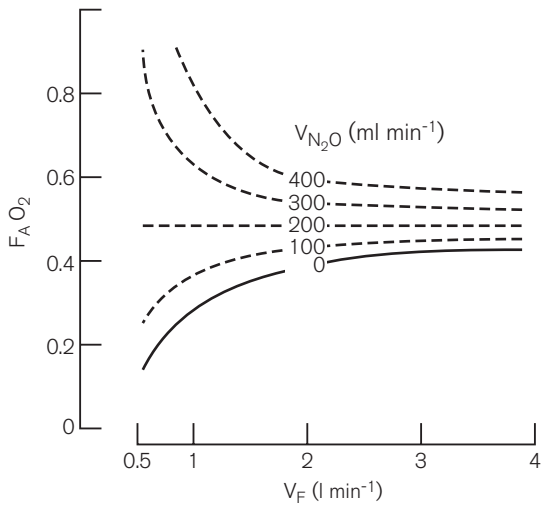


Figure 5.22 Predicted variations in alveolar oxygen concentrations (F_AO_2) produced by decreasing fresh gas flow (V_F) and assuming different levels of nitrous oxide uptake in a 50:50 mixture of oxygen in nitrous oxide. A constant oxygen consumption of 200 ml min^{-1} and a constant ventilation required to produce an alveolar CO_2 concentration of 5% have been assumed. The heavy line shows the fall in alveolar oxygen concentration with decreasing fresh gas flow when the inspired and body levels of nitrous oxide have reached equilibrium (zero uptake).
(From Scurr C, Feldman SA, Soni N, editors. *Scientific Foundations of Anaesthesia*. London: Butterworth/Heinemann; 1990, with permission.)

under controlled conditions. It can be seen that at a FGF of 1 l min^{-1} if there is no longer any nitrous oxide uptake, the oxygen concentration has dropped to 27% and drops even further at a FGF of 0.5 l min^{-1}. Therefore, in clinical practice the oxygen concentration in a circle at low flows is unpredictable and monitoring of inspired oxygen with an analyzer is essential. In fact, monitoring of all gasses and anaesthetic vapours should be considered mandatory for circle systems at low flows.

The use of volatile agents in the circle system

Volatile inhalational agents can be introduced into the system either by being added to the fresh gas flow (vaporizer outside circle/VOC) or by incorporating the vaporizer within the circle (vaporizer in circle/VIC).

Vaporizer outside circle

This is probably the most common method of introducing inhalational agents into the breathing system. At high fresh gas flows, the vapour concentration in the circle is reliably represented by the dial setting of the vaporizer. However, as the FGF is reduced two phenomena occur:

- Expired gas, which is recycled (and which has a reduced concentration of inhalational agent due to uptake), dilutes the FGF within the system. This reduces the delivered concentration of inhalational agent to below that anticipated by the dial setting on the vaporizer.
- At a low FGF, vaporizer efficiency may well be altered, providing a lower or higher than expected concentration of inhalational agent. This is not a problem in modern vaporizer performance but was seen with models that are now obsolete.

At low FGFs, and in the absence of a vapour analyzer, therefore, the anaesthetist would need to know:

- the performance of the vaporizer in use at that given FGF
- the expired concentration of inhalational agent
- the degree of dilution of the FGF with expired gas (which in turn depends on the patient's minute ventilation).

The lower the FGF, the more difficult it is to predict the inspired concentration of agent. At flows below 2 l min^{-1} it is essential to incorporate a vapour analyzer into the system, especially during controlled ventilation when signs of light anaesthesia may be more difficult to determine, to ensure that adequate amounts of agent reach the patient.

Vaporizer in circle

If the vaporizer is incorporated into the circle system, it must have a low resistance to gas flow so as to minimize the respiratory work required of a spontaneously breathing patient. High-resistance plenum vaporizers are unsuitable.

With a vaporizer incorporated in the circle, recirculating gas picks up vapour to add to the vapour that is already being carried, and, therefore, the vapour concentration may well be greater than intended. Calibration of vaporizers in this system is, therefore, impossible. The vapour concentration in this type of system depends on a number of factors when equilibrium has been reached:

- *The fresh gas flow*. The lower the FGF, the greater will be the recirculation of gas already carrying vapour and the higher will be the concentration of inspired agent.
- *The efficiency of the vaporizer*. The more efficient the vaporizer the higher will be the inspired concentration of agent. This has important implications with potent inhalational agents. At a low FGF and a large assisted or controlled minute ventilation, the anaesthetic gas may well become saturated with agent at that temperature. For halothane, this will represent a concentration of 33% at 21°C (the saturated vapour pressure for halothane).

Therefore, for potent agents, an inefficient vaporizer is preferable. The presence of a wick in a vaporizer in the circle is also unsuitable, since water vapour will condense on the wick, reducing its efficiency and possibly increasing the resistance to gas flow.

Ether, for which much higher concentrations are appropriate, has, however, been widely and safely used with a VIC. Adequate vaporization may be assisted by the use of baffles within the vaporizer that cause the gasses to impinge repeatedly on the surface of the ether, or even by bubbling the fresh gas flow through the liquid ether. It may also be increased to some extent by the heat from the recirculating expired gasses.

Breathing systems with assisted circulation

It has long been appreciated that part of the energy required to propel the gasses along the passages of a breathing system must be derived from the patient's respiratory effort. The latter, prejudiced by the depressant effects of narcosis, can become inefficient, and any form of resistance would further impair respiratory function. On the other hand, any form of assistance to flow or of reduction of resistance would be beneficial. To this end various 'circulators' had been devised in the past although none of these has persisted through to modern usage. However, there is renewed interest in this facility and the



Figure 5.24 Double-ended bag for paediatric anaesthesia.

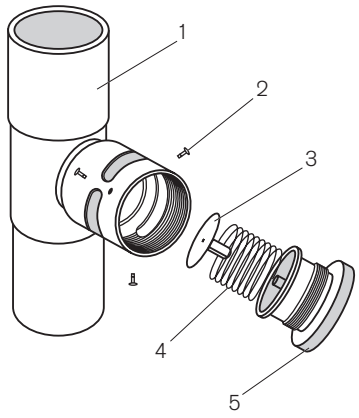


Figure 5.25 The Heidbrink valve: 1, male tapers at both ends; 2, retaining screws; 3, disc; 4, spring; 5, valve top.

part of the connector that fits inside the bag. This prevents the inlet from being occluded if the bag were folded.

In the Jackson Rees T-piece paediatric attachment, a *double-ended* bag is added to the expiratory limb (Fig. 5.24) and the smaller end acts as an expiratory port, the aperture of which can be controlled by the anaesthetist.

The material of which a breathing system bag is constructed is important. Where ventilation is spontaneous, the opening pressure of the expiratory valve must exceed that required to prevent the bag from emptying spontaneously owing to its weight or resistance to distension. Therefore, to maintain a low expiratory pressure, the bag must be 'soft'. This was achieved easily when natural latex rubber bags were in common use. The increase in latex allergy in the general population and in healthcare workers has had a major impact on the provision of equipment that had previously contained natural latex rubber. All anaesthetic equipment in the UK is now supplied with natural latex-free parts. These include ventilator bellows and reservoir bags. The latter are made from synthetic compounds such as polychloroprene (neoprene, a synthetic latex rubber). Some manufacturers make their bags sufficiently pliant to mimic the elasticity of natural latex. These will distend well-beyond their normal filling capacity until they burst, but the pressure in the bags will stay below 60 cm H_20. This is thought to be a safety feature that prevents barotrauma to a patient's lungs should the exhalation pathway in a breathing system containing this bag become occluded. This would be caused most commonly by an APL valve inadvertently screwed shut. Some, however, are made from less compliant material and the pressure may rise to over 60 cm H_20. These bags should be fitted only to breathing systems that have a pressure relief valve attached either as a separate feature or built into the APL valve (see below).

The observed movement of the bag depends on several factors, such as its shape, size, degree of filling, the tension of the expiratory valve and the fresh gas flow rate, as well as on the patient's tidal volume. An accurate estimate of the patient's tidal volume cannot be made simply by watching the bag.

Adjustable pressure limiting (APL) valves

The purpose of this valve is to allow the escape of exhaled (expired) and surplus gasses from a breathing system, but without permitting entry of the outside air, even during a negative phase. Usually it is desirable that the pressure required to open the valve should be as low as possible, in order to minimize resistance to expiration. It must, however, present sufficient resistance to prevent the reservoir bag from emptying spontaneously, particularly when a scavenging system is employed that exerts a slight subatmospheric pressure upon it.

The operating principles of APL valves are based on the Heidbrink valve (Fig. 5.25). The valve disc is as light as possible, and rests on a 'knife-edge' seating that presents a small area of contact. This lessens the tendency to adhesion between the disc and seating due to the surface tension of condensed water from the expired air, or after washing or sterilizing. The disc has a stem that is located in a guide, in order to ensure that it is correctly positioned on the seating, and a lightweight coiled spring, which promotes closure of the valve.

The spring is a delicate coil and is of such dimensions that when the valve top is screwed fully 'open' there is minimal pressure on the disc when seated. However, during the 'blow-off' phase the disc rises and shortens the spring so that the pressure it exerts on the disc is greater and will close it at the appropriate time. Screwing down the valve top produces progressively increasing tension in the spring. When the top is screwed down fully, the valve is completely closed. The valve should always be used in the vertical position so that the disc is seated by gravity.

The Humphrey APL valve

An interesting addition to the standard APL valve is the modification seen on the version (Figs 5.13 and 5.26) that is part of the Humphrey ADE system. Here, the valve disk is attached to a red-coloured spindle that extends through the valve top. When the valve is fully open, the spindle is seen to bob up and down as the disk is lifted up and down during respiration. The valve top is made concave and shiny so that it reflects and magnifies the spindle colour so as to detect even the smallest movement when used in paediatric anaesthesia. The inside of the valve body has a small funnel through which the disk has to move before significant gas can escape. This initial movement of 5 mm accentuates the bobbing action of the spindle which is useful particularly in paediatric anaesthesia.

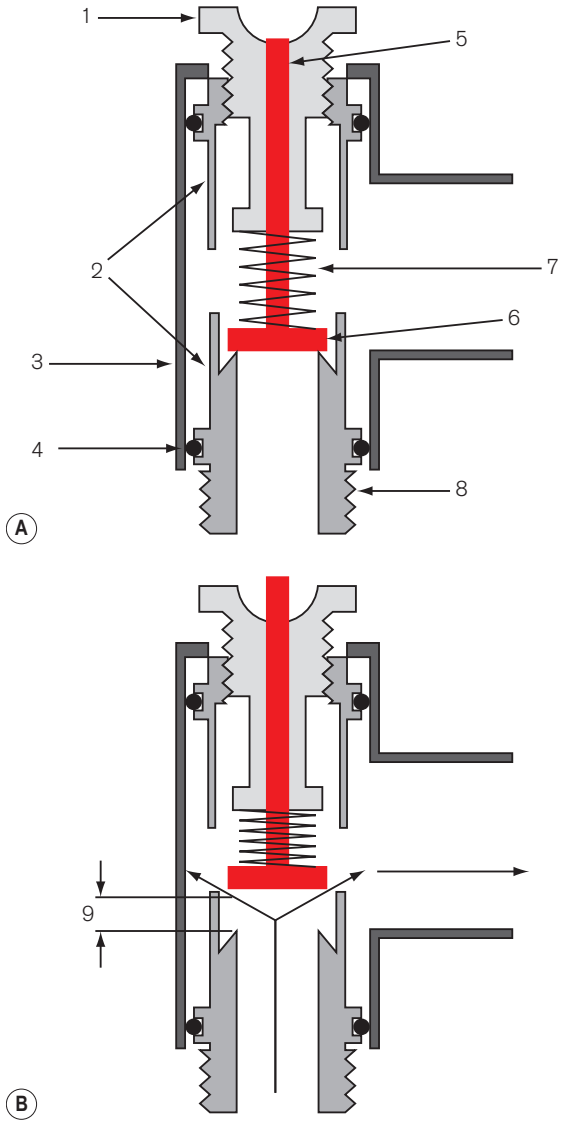


Figure 5.26 'Humphrey' APL valve. **A.** Inspiration (with valve disc closed); **B.** beginning of exhalation (valve disc just clearing small funnel). (1) Valve top; (2) valve body; (3) exhaled gas scavenging shroud; (4) 'O'-ring seal between the valve body and shroud; (5) valve spindle; (6) valve disc; (7) valve spring; (8) screw thread to secure valve to the breathing system; (9) 5 mm funnel which accentuates movement of the spindle.

When the ADE system is used in the Mapleson 'A' mode, the valve spindle may be held down with a finger if switching from spontaneous to manual ventilation is required. This has a number of advantages:

- As there is no leak from the valve during inspiration, the patient's lung compliance may be more accurately assessed.
- It becomes as efficient as the Mapleson 'A' mode in spontaneous respiration (i.e. the valve is shut during inspiration and open during expiration).
- The valve top may be kept in the 'open' position at all times and does not require repeated adjustment when switching between spontaneous and manual ventilation.

It was originally thought that the increased respiratory work produced by the expiratory resistance of APL valves was detrimental to anaesthetized patients. This is without doubt true when the valve resistance is high (due to sticky valves, narrow valve apertures) or where the respiratory effort is severely compromised (e.g. in neonates). However, modern valve design (with wider valve apertures, lighter valve discs, more delicate springs, better screw threads) minimizes this resistance. Furthermore a small PEEP (positive end expiratory pressure) effect that these valves may produce is now thought to be positively beneficial, reducing the potential for the functional residual capacity of the lungs to fall below the closing volume in supine anaesthetized patients.

APL valves with in-built overpressure safety devices

Any unexpected pressure rise in a breathing system was made relatively safe by the compliance of the latex rubber of the reservoir bag. This could still fail, for instance if the bag were trapped under the wheel of an anaesthetic machine, a dangerously high pressure could develop within the breathing system and be passed on to the patient. Now that latex rubber is no longer used owing to the increase in latex allergy, new materials are used that may not have the same elasticity and so are not as compliant. This safety feature can no longer be relied on. APL valves are now available in which an overpressure safety device has been incorporated.

An example, is shown in Figure 5.27.

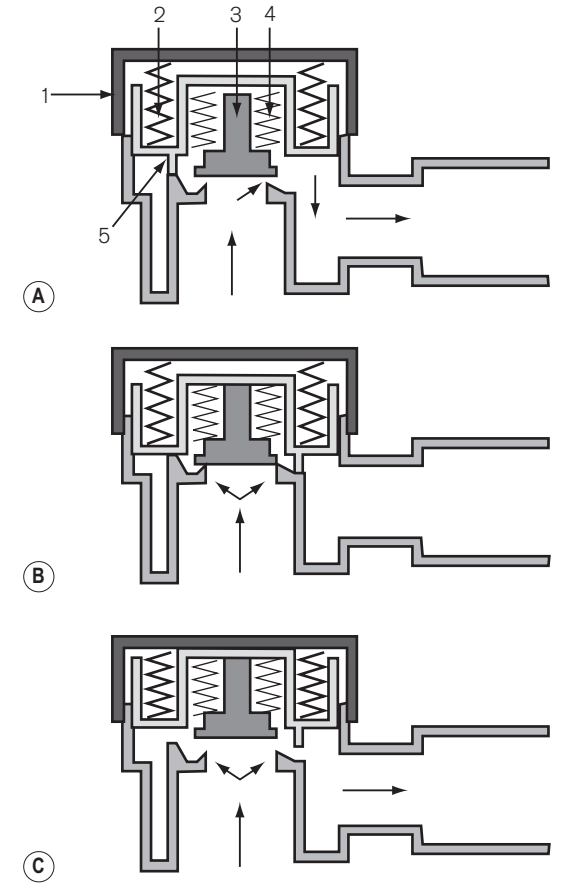




Figure 5.27 An APL valve with an overpressure safety device. (1) Valve control knob; (2) high-pressure spring; (3) valve spindle; (4) light-pressure spring; (5) the part of the asymmetric valve body that rotates during valve closure and occludes the expiratory limb. The figure shows **A.** the valve open during exhalation; **B.** the valve closed; and **C.** the valve closed with the overpressure safety device in action. **D.** Dräger calibrated APL valve with overpressure relief valve.

It has two valves:

- the inner (3), which is tensioned with a weak spring (4)
- the outer (5), which is tensioned with a more powerful one (2).

When the valve top (1) is unscrewed fully (Fig. 5.27A), the outer valve is permanently open, but the inner one is closed until exhaled gas forces it open. The pressure required to do this is small (0.15 kPa /1.5 cm H_2O). As the valve is gradually closed, the expiratory flow resistance increases and in sophisticated examples this resistance can be calibrated and displayed on the valve body. When the valve top is screwed down fully, both valves are closed and in this position the outer one is pushed against the inner so that is has no movement of its own (Fig. 5.27B). An excess pressure is now required to move the more powerful spring on the outer valve which will begin to open at 3 kPa (30 cm H_2O) and be fully open between 6 and 7 kPa (60–70 cm H_2O) when the gas flow is 50 l min^{-1} (Fig. 5.27C). Fig. 5.27D is an example of a calibrated APL valve incorporating an overpressure safety valve set at 70 cm H_2O.

Alternative APL valve design

Many breathing systems are now supplied as single use items. This includes the APL valve (Fig. 5.28A). It is now possible to simplify the design so that the spring and the valve disc are replaced by a neoprene flap valve (1) (Fig. 5.28Bi). This opens and shuts in the normal manner during spontaneous respiration. When positive pressure is required, the valve top (2) operates a screw threaded insert (3) that lowers an overpressure relief valve (4). As the valve top is screwed shut, the insert lowers the second valve (4) onto the flap valve housing, gradually occluding the expiratory pathway until complete occlusion occurs (Fig. 5.28Bii). The second valve is fitted with a spring (5) which is strong enough to maintain the occlusion up to a pressure of 60 cm H_2O, but weak enough for the valve to lift (Fig. 5.28Biii) and allow gasses at a higher pressure to escape.

Breathing hoses

The hoses connecting the components of a breathing system must be of such a diameter as to present a low resistance to gas flow. Its cross-section must be uniform to promote laminar flow where possible and, although it should be flexible, kinking should not occur.

The most commonly used type was for a long-time, corrugated hose of rubber or polychloroprene (neoprene). The corrugations allow acute angulations of the hose without kinking. The advantage of these materials is that the ends are more easily stretched, and will make a good union with other components of different diameters. They



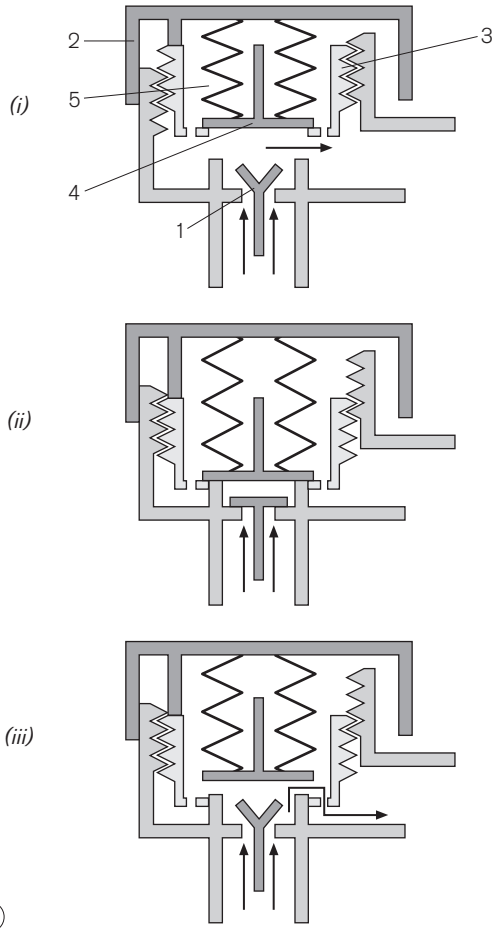


Figure 5.28 **A.** A single-use APL valve (Intersurgical). **B.** Operation of an Intersurgical single use APL valve. (1) Neoprene flap valve; (2) valve top; (3) screw threaded insert; (4) overpressure relief valve; (5) spring. (i) Valve in open position. (ii) Valve shut. (iii) Overpressure relief valve operated.

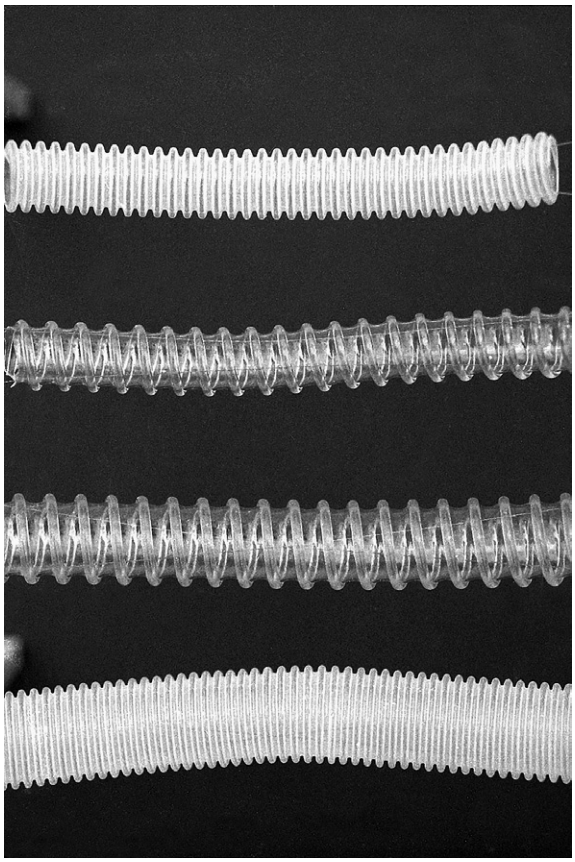


Figure 5.29 From top to bottom: 15 mm corrugated plastic breathing hose; 15 mm smooth-bore hose; 22 mm smooth-bore hose; 22 mm corrugated hose.

may be sterilized by steam autoclaving and be reused in countries where single-use alternatives are impractical. The disadvantages are that the irregular wall must cause turbulence and being opaque may harbour dirt and infection unseen. They are also heavy and, if unsupported, may drag on a facemask or endotracheal tube.

Various other materials such as silicone rubber and plastics (polyethylene) are in use, both in corrugated and smooth form (Fig. 5.29). Smooth bore breathing hose produces less turbulence than the corrugated variety at similar gas flows. It can also be produced so that it resists kinking (by the attachment of a reinforcing spiral of a similar material to its external surface). With smooth bore hosing, a smaller diameter (e.g. 15 mm) may well be acceptable for use with adult breathing systems (see Humphrey ADE breathing system).

Plastic hosing has become very popular because it is lightweight, cheap to manufacture, and, therefore, disposable. Some are supplied as complete breathing systems

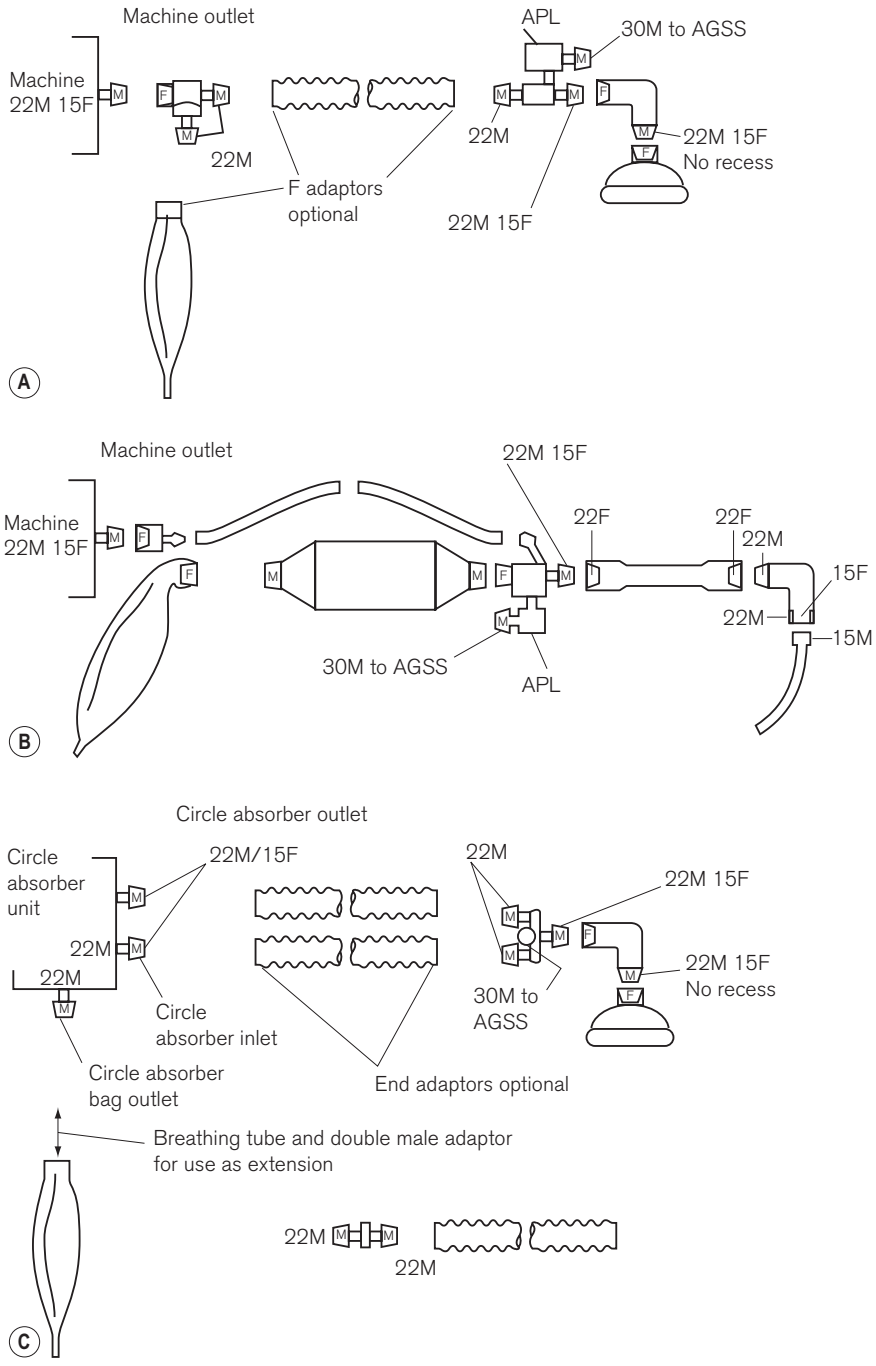


Figure 5.32 Current sequence of tapers in anaesthetic breathing systems as recommended by the ISO and BSI (BS 3849) as from 1988 and maintained in BS EN ISO 5356-1: 2004. Typical layout of **A.** Mapleson A system; **B.** to-and-fro absorption system; and **C.** circle absorption system. M, male conical fitting; F, female conical fitting; APL, adjustable pressure-limiting valve.

ends. One will fit the bag mount (male to female) but the other will not fit the bag (female to female). A male to male 22 mm tapered adapter is required (see Fig. 5.31).

Problems with tapered connections

Many accidents have occurred as a result of the accidental and unobserved disconnection of tapered joints, especially between the breathing hose and other components of the breathing system such as catheter mounts. The material used in the construction may either wear with frequent use (most plastics and rubber) or become distorted by damage (metal connectors). All taper fit connections must be assumed to be prone to accidental disconnection. Disconnection can be minimized by giving the components of the joint a slight twist following their insertion.

Conversely some metal connectors made from aluminium alloys may stick together by the phenomenon of cold welding produced by the recommended twist above, and may be very difficult to separate.

Reuse of breathing system components

Items that are intended for *reuse* (i.e. to be used on different patients) are normally made from materials (rubber, neoprene, silicone and metal) that withstand repeated autoclaving.

Some items are designated as *single use* (i.e. to be used on one person only) if they are made from materials such as plastics that are not easily sterilized. Also, if the tapers (see above) are made from materials that are easily distorted with repeated connection they will be designated for single use only.

A third category has been described. Some items that might be designated as single use because they are not easily sterilized may be acceptable for *repeated use*. Provided that these items are protected by a high-quality bacterial filter, some manufacturers will accept product liability if the item is used on a number of patients as long as the item is thoroughly checked between uses.

FURTHER READING

Breathing systems

Boulton TB. Breathing systems. Editorial. *Anaesthesia* 1979;**34**:605–7.

Classification of breathing systems

Cook LB. Mapleson breathing systems. The importance of the expiratory pause. *Anaesthesia* 1996;**51**:453–60.

Dorrington KL.The Mapleson breathing systems. *Anaesthesia* 1996;**51**:988.

Mapleson WW. The elimination of rebreathing in various semi-closed anaesthetic systems: 1954. *Br J Anaesth* 1998;**80**:263–9.

Mapleson WW. Editorial I: Fifty years after – reflections on 'The elimination of rebreathing in various semi-closed anaesthetic systems'. *Br J Anaesth* 2004;**93**:319–21.

Miller DM. Breathing systems reclassified. *Anaesth Intensive Care* 1995;**23**:281–3.8.

Mapleson A systems

Chan AS, Bruce WE, Soni N. A comparison of anaesthetic breathing systems during spontaneous ventilation. An in-vitro study using a lung model. *Anaesthesia* 1989;**44**: 194–9.

Ooi R, Pattison J, Soni N. The additional work of breathing imposed by Mapleson A systems. *Anaesthesia* 1993;**48**:599–603.

Ooi R, Lack A Soni N, Whittle J, Pattison J. The parallel Lack anaesthetic breathing system. *Anaesthesia* 1993;**48**:409–14.

Mapleson D, E and F systems

Adams AP. The Bain circuit. Prevention of anaesthetic mixture dilution when using mechanical ventilators delivering non-anaesthetic gasses. *Anaesthesia* 1977;**32**:46–9.

Henville ID, Adams AP. The Bain anaesthetic system. *Anaesthesia* 1976;**31**:247–56.

Spoerel WE, Bain JA. Anaesthetic breathing systems. *Br J Anaesth* 1986;**58**:819–21.

Rebreathing with Mapleson systems

Barrie JR, Beatty PC. Rebreathing and semiclosed anaesthetic breathing systems. *Anaesthesia* 1993;**48**:86–7.

Cook LB. Rebreathing in the Mapleson A, C and D breathing systems with sinusoidal and exponential flow waveforms. *Anaesthesia* 1997;**52**: 1182–94.

Dorrington KL, Lehane IR. Rebreathing during spontaneous and controlled ventilation with 'T' piece breathing systems: a general solution. *Anaesthesia* 1989;**44**:300–2.

Dorrington KL, Lehane JR. Minimum fresh gas flow requirements of anaesthetic breathing systems during spontaneous ventilation: a graphical approach. *Anaesthesia* 1987;**42**: 732–7.

Resistance to flow with Mapleson systems

Kay B, Beatty PC, Healy TE, Accoush ME, Calpin M. Change in the work of breathing imposed by five anaesthetic breathing systems. *Br J Anaesth* 1983;**55**:1239–47.

Martin DG, Kong KL, Lewis GT. Resistance to airflow in anaesthetic breathing systems. *Br J Anaesth* 1989;**62**:456–61.

Hybrid systems

Humphrey D. A new anaesthetic breathing system combining Mapleson A, D & E principles. *Anaesthesia* 1983;**38**:361–72.

Humphrey D. The ADE anaesthetic breathing system. *Anaesthesia* 1984;**39**:715–6.

Humphrey D, Brock-Utne JG, Downing JW. Single lever Humphrey ADE lowflow universal anaesthetic breathing system.

Part II: Comparison with Bain system in anaesthesized adults during controlled ventilation. *Can Anaesth Soc J* 1986;**33**: 710–8.

Orlikowski CEP, Ewart MC, Bingham RM. The Humphrey ADE system: evaluation in paediatric use. *Br J Anaesth* 1991;**66**:253–7.

Carbon dioxide absorption to and fro systems

Shaw M, Scott DH. Performance characteristics of a 'to and fro' disposable soda lime canister. *Anaesthesia* 1998;**53**:454–60.

Circle systems

Bracken A, Cox LA. Apparatus for carbon dioxide absorption. *Br J Anaesth* 1968;**40**:660–5.

Levels of anaesthetic agents in circle systems

Conway CM. Gaseous homeostasis and the circle system. Validation of a model. *Br J Anaesth* 1986;**58**: 337–44.

Cook LB, Chakrabarti MK. Circle systems with a coaxial inspiratory limb. Investigation with a lung model. *Anaesthesia* 1996;**51**: 247–54.

Mapleson WW. The concentration of anaesthetics in closed circuits, with special reference to halothane. *Br J Anaesth* 1960;**32**:298–309.

Mapleson WW. The theoretical ideal fresh-gas flow sequence at the start of low-flow anaesthesia. *Anaesthesia* 1998;**53**:264–72.

Sobreira DP, Jreige MM, Saraiva R. The fresh-gas flow sequence at the start of low-flow anaesthesia. *Anaesthesia* 2001;**56**:379–80.

Sobolev I, Sellers WF. Rate of change in gas concentrations in a charged circle system with absorber. *Anaesthesia* 2001;**56**:380–1.

Wright D, Brosnan S, Royston B, White D. Controlled ventilation using isoflurane with an in-circle vaporiser. *Anaesthesia* 1998;**53**:650–3.

Absorbers and absorbents

Coetzee JF, Stewart LJ. Fresh gas flow is not the only determinant of volatile agent consumption: a multi-centre study of low-flow anaesthesia. *Br J Anaesth* 2002;**88**:46–55.

Cossham PS. Obstruction to wet soda lime granules. *Anaesthesia* 1992;**47**: 10–1.

Knolle E, Linert W, Gilly H. The color change in CO2 absorbents on drying: an in vitro study using moisture analysis. *Anesth Analg* 2003;**97**:151–5.

Murray JM, Renfrew CW, Bedi A, McCrystal CB, Jones DS, Fee JP. Amsorb: a new carbon dioxide absorbent for use in anesthetic breathing systems. *Anesthesiology* 1999;**91**:1342–8.

Carbon monoxide and compound A formation

Baum JA, Woehlckhemical JH. Reaction interaction of inhalation anaesthetics with CO_2 absorbents. *Best Pract Res Clin Anaesthesiol* 2003;**17**:63–76.

Baxter PJ, Garton K, Kharasch ED. Mechanistic aspects of carbon monoxide formation from volatile anesthetics. *Anesthesiology* 1998;**89**:929–41.

Cunningham DD, Huang S, Webster J, Mayoral J, Grabenkort RW. Sevoflurane degradation to compound A in anaesthesia breathing systems. *Br J Anaesth* 1996;**77**:537–43.

Fang ZX, Eger EI 2nd. USCF Research shows that CO comes from CO2 absorbent. *Anaesth Patient Saf Found Newsl* 1995;**9**:26–9.

Fang ZX, Eger EI 2nd, Laster MJ, Chortkoff BS, Kandel L, Ionescu P. Carbon monoxide production from degradation of desflurane, enflurane, isoflurane, halothane, and sevoflurane by soda lime and Baralyme. *Anesth Analg* 1995;**80**:1187–93.

Funk W, Gruber M, Wild K, Hobbhahn J. Dry soda lime markedly degrades sevoflurane during simulated inhalation induction. *Br J Anaesth* 1999;**82**:193–8.

Harrison N, Knowles AC, Welchew EA. Carbon monoxide within circle systems. *Anaesthesia* 1996;**51**: 1037–40.

Knolle E, Heinze G, Hermann G. Small carbon monoxide formation in absorbents does not correlate with small carbon dioxide absorption. *Anesth Analg* 2002;**95**:650–5.

Knolle E, Linert W, Gilly H. The color change in CO2 absorbents on drying: an in vitro study using moisture analysis. *Anesth Analg* 2003;**97**:151–5.

References Medicines Control Agency Drug Alert 1995 (N31 May EL(95) (ALERT) A/17) Important precautions required when using halogenated anaesthetic agents. 1995;**78**:340–8.

Stabernack CR, Brown R, Laster MJ, Dudziak R, Eger EI 2nd. Absorbents differ enormously in their capacity to produce compound A and carbon monoxide. *Anesth Analg* 2000;**90**:1428–35.

Strum DP, Edmond I, Eger II. Degradation, absorption and solubility of volatile anaesthetics in soda lime depend on water content. *Anesth Analg* 1994;**78**: 340–8.

Struys MM, Bouche MP, Rolly G, Vandevivere YD, Dyzers D, Goeteyn W et al. Production of compound A and carbon monoxide in circle systems: an in vitro comparison of two carbon dioxide absorbents. *Anaesthesia* 2004;**59**:584–9.

Woehlck HJ, Dunning MB 3rd, Kulier AH, Sasse FJ, Nithipataikom K, Henry DW. The response of anesthetic agent monitors to trifluoromethane warns of the presence of carbon monoxide from anesthetic breakdown. *J Clin Monit* 1997;**13**:149–55.

Yamakage M, Kimura A, Chen X, Tsujiguchi N, Kamada Y, Namiki A. Production of compound A under low-flow anesthesia is affected by type of anesthetic machine. *Can J Anaesth* 2001;**48**:435–8.

Abnormal heat generation in circle systems

Castro BA, Freedman LA, Craig WL, Lynch C. Explosion within an anesthesia machine: Baralyme(R), high fresh gas flows and sevoflurane concentration. *Anesthesiology* 2004;**101**:537–9.

Fatheree RS, Leighton BL. Respiratory distress syndrome after an exothermic Baralyme(R)–sevoflurane reaction. *Anesthesiology* 2004;**101**: 531–3.

Laster M, Roth P, Eger EI 2nd. Fires from the interaction of anesthetics with desiccated absorbent. *Anesth Analg* 2004;**99**:769–74.

Wu J, Previte JP, Adler E, Myers T, Ball J, Gunter JB. Spontaneous ignition, explosion, and fire with sevoflurane and barium hydroxide lime. *Anesthesiology* 2004;**101**:534–7.

Apparatus

Blanshard HJ, Milne MR. Latex-free reservoir bags: exchanging one potential hazard for another. *Anaesthesia* 2004;**59**:177–9.

Department of Health. Anaesthetic and respiratory equipment: the use of 22 mm breathing system connections. *Safety Action Bulletin* 1989.

Medical Devices Agency (MDA). DB 9501 Reuse of Medical devices supplied for single use only. *Reuse of Equipment Intersurgical Technical Bulletin TB* 1995; **5**(4):95.

Chapter 6

Airway management equipment

Tim Cook

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Standards, techniques (and fashions) in airway management have changed considerably over the last two decades. Although the most basic principle of anaesthesia is the maintenance of a patent airway aimed at providing adequate oxygenation and ventilation, the range of devices and products used has seen major changes especially in the focus of the designs. Of the developments in the last decade several are notable.

The laryngeal mask in its various forms has continued to expand both within and beyond routine airway management. Prior to the introduction of what is now branded as the LMA Classic, there was usually a clear choice between tracheal intubation and facemask application, each with its own distinct advantages and problems. The laryngeal mask is now the commonest device for airway management during anaesthesia in the UK and is used for approximately 56% of cases.

Numerous alternatives to the LMA have since been developed. These devices which sit outside the larynx and aim to provide a gas-tight seal are now generally referred to simply as *supraglottic airways* or *SADs* (supraglottic airway devices): the term *extraglottic airway* is more anatomically accurate but is not widely used.[1] The number and variety of SAD designs have expanded dramatically in the last few years with the result that there are numerous products that vary considerably in materials and performance. Many of these are single-patient use laryngeal masks designed to compete for the market share of the original LMA. Nonetheless several newer SADs offer genuine advances in versatility, efficacy and safety.

The use of the flexible fibreoptic endoscope (fibrescope) for intubation continues to increase and to emerge from being the domain of a few to becoming a mainstream activity at which anaesthetic trainees rightly feel they must be adept. Technological developments are apparent here too with an increasing shift towards miniaturized digital video-based devices rather than optical glass-fibre image transmission.

In the last few years, we have also seen the proliferation of a new generation of crossover devices for laryngoscopy which combine features of both fibrescopes and traditional rigid (lighted retractor type) laryngoscopes. These rigid devices with predefined shapes, rely instead on fibreoptics, prisms and mirrors or digital camera technology for image transmission from the tip of the device. The standard technique of direct laryngoscopy for tracheal intubation is being challenged. While some commentators suggest that direct laryngoscopy has a limited future, the precise role of 'indirect' laryngoscopy with such devices in elective and emergency procedures, and in cases of airway difficulty is far from established. It is certain that the development and introduction of these devices is currently outstripping the profession's capacity (or desire) to carefully evaluate their proposed benefits. Unanticipated complications may accompany some techniques, and harm, albeit rare, may be seen with this technology too.

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Finally, overlying all the above considerations; concern about the transmission of infective agents, particularly those responsible for variant Creutzfeldt–Jakob disease (vCJD), has been the apparent driving force behind an increasing trend towards reliance on single-use airway devices. This trend, whether based on science or not (of which more later) has meant that the majority of airway devices now being developed are at least in part single-use. Production and material costs and hence price, as well as storage and stocking issues, are ultimately more significant considerations for single-use items and this is already affecting the range and quality of equipment available in many hospitals.

This chapter addresses all these areas: it will be evident that the field of airway management has seen a massive proliferation of devices and techniques. Whereas it was once possible for this book to be a complete and up-to-date inventory of all equipment that anaesthetists were likely to come across in their professional lives, our more modest aim now, particularly in this chapter, is to represent examples and classes of devices that are currently in popular use and to further illustrate important themes in such developments.

FUNDAMENTALS

The fundamental principle of airway management during general anaesthesia is that:

1. a *clear airway* must be established and maintained allowing free passage of gasses to and from the patient's lungs. Failure to do this is likely to lead to collapse of the airway at pharyngeal or laryngeal level. Without intervention airway obstruction is inevitable and if not correctly managed, will be followed within minutes by hypoxia, brain damage and death.

Additionally it may be necessary to provide:

2. a *sealed airway* that enables controlled (positive pressure) ventilation and prevents loss or dilution of anaesthetic gasses
3. a *protected airway* that prevents the respiratory system being contaminated by regurgitated gastric contents or pharyngeal debris.

These principles, which are not new, encompass the prime aims of airway management and constitute one pillar of anaesthetic practice: safe and effective oxygenation and pulmonary ventilation.

MATERIALS USED IN AIRWAY DEVICES

As ever, the choice of material is governed by the ideal characteristics required for the intended function; taking into account patient and environmental safety and attendant legislation, balanced against the costs of the raw material and production costs.

Rubber

The term 'rubber' accurately refers to the product extracted from the rubber plant, but is now used generically to describe an elastic solid and may be prefixed by a more specific adjective to describe the material used in its production. *Natural latex rubber* is extracted from the bark sap of rubber plants as the monomer C_5H_8. It is washed and then polymerized using acetic acid into its final form for the manufacture of gloves and other highly elastic products. Although cheap and versatile, it degrades easily and is contaminated by plant proteins that are the cause of allergic reactions in susceptible individuals. This problem may be reduced by vigorous washing in the early stages of production to remove the contaminant allergens. Natural latex may be converted to red/black rubber by a process known as vulcanization, where high temperatures destroy the allergenic proteins. *Synthetic latex* materials (nitrile, butyl rubber, and polychloroprene, more commonly known as neoprene) are similar to natural latex, but being synthetic have a lesser allergenic potential. *Silicone rubber* is another synthetic compound not related to latex polymers. All these materials can withstand sufficiently high temperatures to be steam sterilized (autoclaved).

Plastics

The most common plastics are polyethylene and polyvinyl chloride (PVC). Polyethylene is cheap, fairly pliable, non-allergenic and can be moulded easily. It is ideal for breathing hoses, connectors and hypodermic syringes but not for the manufacture of reusable items as it will not withstand autoclaving and components may deform with repeated use. It is unsuitable for tracheal tubes (see below) as other materials cannot be easily welded or glued to its surface (e.g. cuffs). For this purpose, PVC is more suitable. PVC is brittle on its own but can be made in varying degrees of softness by the addition of plasticiszers (see below). Concerns have been raised in some areas about potentially toxic chemicals in PVC called phthalates and the potential production of both phthalates and dioxins when PVC is disposed of or incinerated.[2]

Materials used in tracheal tube construction are described in more detail below (p.156)

ARTIFICIAL AIRWAYS

Terminology

The term *airway* (or *anatomical airway*) will be used to describe the air passages within the subject, including and

beyond the nasal and oral openings. The term *distal* refers to the part of the airway or device furthest into the subject's airway and the term *proximal* to the part emerging or closest to the mouth or nose. *Artificial airway* then is any device that aims to maintain patency of any of the air passages. Artificial airways may be:

- *simple airway adjuncts*, such as the oropharyngeal and the nasopharyngeal airways. These may not be sufficient to maintain the patency of the airway on their own and may require the patient's jaw to be supported as well. Although these are strictly speaking supraglottic devices in that their distal orifice lies above the glottis, in general usage the term SAD is not used to encompass or imply these simple airways.
- *supraglottic airway devices* (SAD) such as the laryngeal mask airway (LMA) and many others which aim to maintain patency of the airway on their own
- *infraglottic devices,* which are airway devices whose distal part lies below the glottis. Therefore they include all tracheal and tracheostomy tubes, jet ventilation catheters and cannulae, and bronchial blockers. It is a lesser used term.

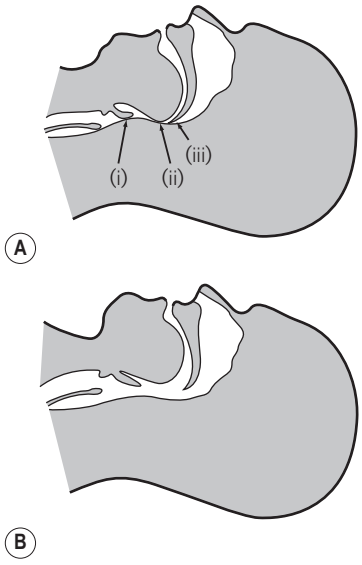


Figure 6.1 **A.** The obstructed airway following the administration of a general anaesthetic if the jaw is unsupported. The airway may be obstructed by (i) epiglottis and/or (ii) tongue pressing on the posterior pharyngeal wall, or by (iii) the soft palate, occluding the oral or nasal airways. **B.** The effect of simple elevation of the jaw in a patient with normal anatomy.

SIMPLE AIRWAY ADJUNCTS

The maintenance of a clear airway in an anaesthetized patient can often be achieved by a simple elevation of the jaw (jaw thrust) and/or extension of the head on the cervical spine. These movements tend to separate the tongue, epiglottis and soft palate from one another and away from the posterior pharyngeal wall (Fig. 6.1). However, in many patients maintenance of the airway in this manner is either ineffective or impractical for surgery. Patients with anatomical reasons for such manoeuvres to fail include those whose 'pharyngeal spaces' are absolutely or relatively small (e.g. a large tongue, small lower jaw, large diameter neck, obesity and also those with large adenoid or palatine tonsils). In such patients the obstruction must be relieved and the easiest way to achieve this is by inserting a device that separates these structures and thus creates an artificial airway. These *airway adjuncts* may be inserted via the mouth (oropharyngeal airway) or via the nose (nasopharyngeal airway) (Fig. 6.2).

Oropharyngeal airway

These devices are shaped to emulate and so restore the space present in the pharynx during consciousness by pushing the tongue and epiglottis away from the posterior pharyngeal wall. Oropharyngeal airways are usually oval-shaped, occasionally circular, in cross-section and are produced in varying lengths and diameters to suit different patient sizes from premature neonate to large adult. The proximal end has a flange to limit the depth of insertion

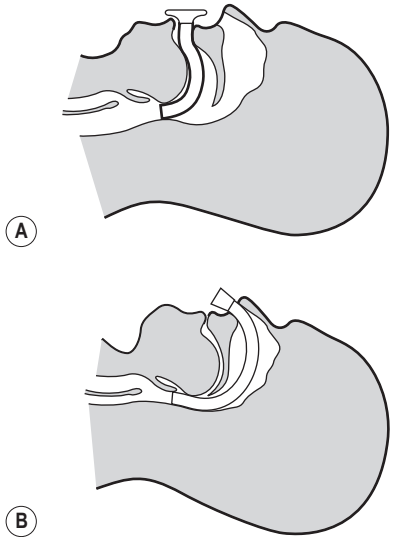


Figure 6.2 **A.** An unobstructed air passage produced with an oropharyngeal airway. **B.** An unobstructed air passage produced with a nasopharyngeal airway.

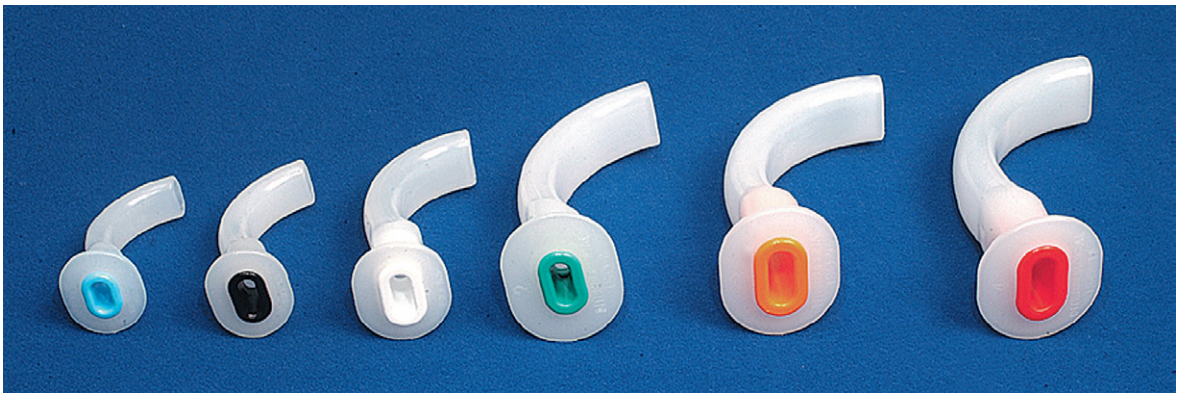


Figure 6.3 Guedel airways in sizes 00, 0, 1, 2, 3, 4 (a smaller size 000 is not shown here).

and prevent its loss into the pharynx. Devices are sufficiently rigid to prevent collapse should the patient bite down and indeed they may be used as a 'bite block' in an intubated patient (to prevent airway obstruction as a result of the tracheal tube being obstructed by biting). When inserted, the distal end should lie at the posterior of the tongue but above the epiglottis. Insertion too deep may in itself lead to airway obstruction either by mechanically pushing the epiglottis back or by irritating the sensitive laryngeal inlet leading to laryngospasm. There is a standard colour and number coding for size. The most popular oropharyngeal airway type is the Guedel pattern (Fig. 6.3). To select the correct size of Guedel airway, the distance from the flange to the distal tip of the airway should be about the same as from the patient's lips to the tragus of the ear.

Inserting the airway

The head is extended, the patient's mouth is opened wide, the lubricated device is inserted so that its concave curvature faces cephalad and its tip is passed along the hard palate. When the tip of the airway reaches the soft palate it is rotated though 180° and the device is advanced in to the pharynx. Alternatively the device can be inserted with its curve in the anatomical position, ideally with a spatula retracting the tongue. There may be resistance to insertion about three-quarters of the way in, which is usually overcome by lifting the angles of the lower jaw forward with the middle fingers of both hands and gently pushing in the flange with both thumbs.

Even when a correctly sized oropharyngeal airway is inserted, airway obstruction may not be resolved. Correct sizing and full insertion increases success, but if obstruction persists alternative sizes should be tried. The airway may be partially pushed out if the patient's jaw is left unsupported and allowed to fall backwards. Chin lift in patients with marked overbite of the upper teeth may also be problematic with the lower teeth acting as a fulcrum so that downward the pressure from the upper teeth tends to dislodge the airway, with risk of partial or complete airway obstruction.

Complications

Gagging, retching or laryngospasm can occur if the airway is inserted in a patient whose pharyngeal reflexes have not been sufficiently depressed by topical or general anaesthesia.

The airway may damage the front teeth, especially if:

- teeth are crowned or part of bridgework
- excessive pressure is exerted due to postoperative masseter spasm or involuntary biting
- there is overenthusiastic jaw support.

Dental damage may also occur without an oropharyngeal airway in place. As dental injury accounts for a substantial proportion of legal claims against anaesthetists, care should be taken to prevent this complication during oropharyngeal airway insertion, removal and when it is in place. Use of a rubber dental prop or *bite block* (see below) between the patient's molar teeth may prevent damage to the incisors caused by the above mechanisms and some would recommend its use during emergence from anaesthesia if a tracheal tube or laryngeal mask airway is in place.

Oropharyngeal airways for flexible endoscopic oral intubation

A number of airways have been devised to assist oral intubation. When used to aid flexible endoscopic intubation they serve to deliver the endoscope behind the tongue, and as close to the larynx as possible, ideally having bypassed any secretions. Their shapes are broadly similar to the Guedel airway but often describing a fuller curvature along the length and with a more circular cross-section (Fig. 6.4).

The Berman airway (originally designed to assist blind oral intubation and pharyngeal suction), like the VBM and Ovassapian (not pictured) airways is open along one side. These devices when used for endoscopic intubation are

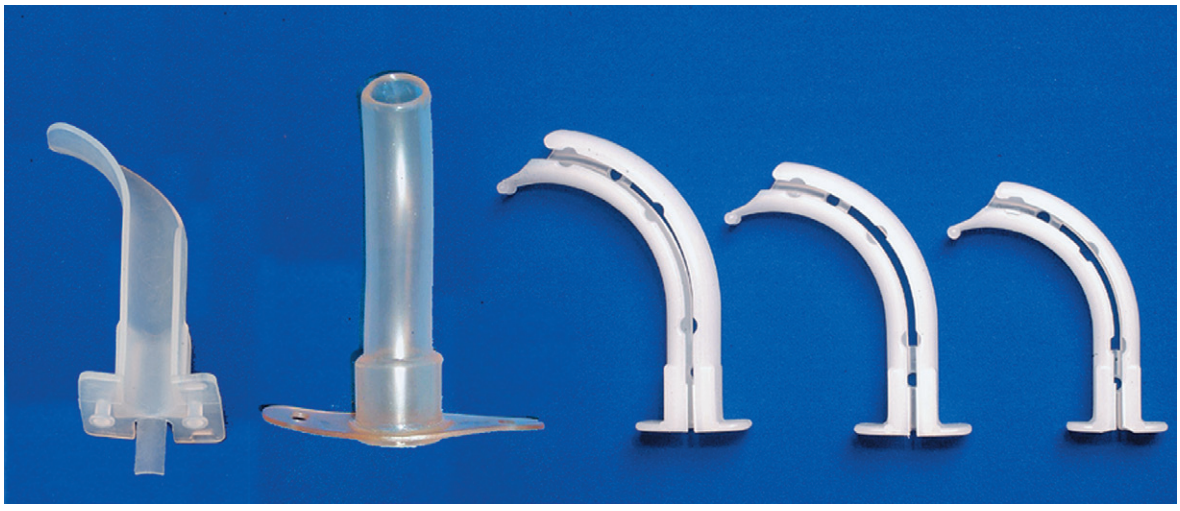


Figure 6.4 (From left to right): VBM intubating airway and bite block, Optosafe airway, Berman airways in three sizes.

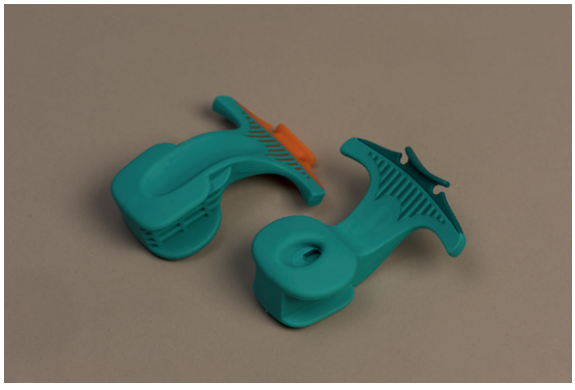


Figure 6.5 BreatheSafe (OGM Ltd, UK) bite block, shown in size 2 (green tag) and 3 (orange tag).

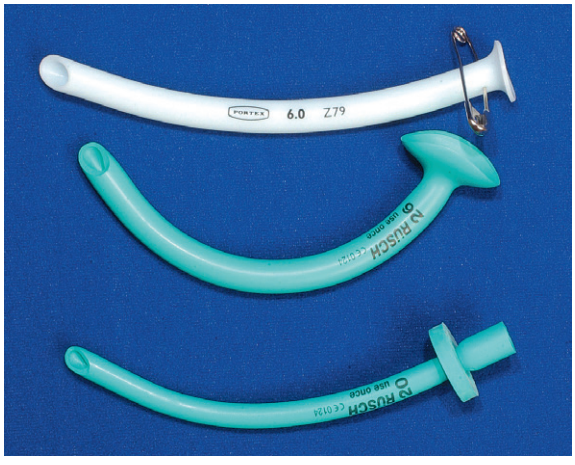


Figure 6.6 Nasopharyngeal airways (from top to bottom) Portex and Rusch designs, the lower with an adjustable flange.

removed before passage of the tracheal tube. The fully enclosed Optosafe airway has a large enough diameter to accommodate a tracheal tube.

The airways also act as a 'bite block' preventing damage to expensive fingers or delicate endoscopes (see also: Flexible endoscopes, conduit airways). This purpose is better served by the use of a dedicated bite block such as the BreatheSafe (Fig. 6.5) which more resembles a dental prop and allows full access to the oral cavity while preventing mouth closure.

All the above devices are available in a number of sizes and correct size selection is important for good function.

Nasopharyngeal airway

The nasopharyngeal airway is designed to be passed through the nares and along the floor of the nose to deliver the tip beyond the soft palate to lie in the oropharynx, above the epiglottis. It can bypass nasal, soft palate and tongue base obstruction. The tubes have either a fixed or adjustable flange (Fig. 6.6) at the proximal end to prevent loss of the device into the nose and to limit insertion of an excessive length. The tip is bevelled to make its passage through the nose less traumatic. Some designs have a hole cut in the wall opposite the bevel to maintain patency if the tip becomes obstructed. They are produced in a range of sizes with the length of the airway governed by the internal diameter of the tube.

Nasopharyngeal airways are often indicated where a patient has limited jaw opening, awkward or fragile dentition or where an oropharyngeal airway is frequently displaced by a marked overbite. They are well tolerated by

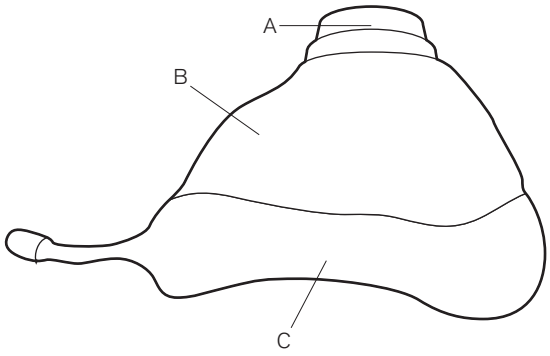


Figure 6.7 The parts of an anaesthetic facemask: A, the mount; B, the body; C, the edge.

patients during relatively light levels of anaesthesia and during emergence. To avoid traumatic insertion and heavy bleeding nasopharyngeal airways are ideally soft (plastic, polyurethane or latex rubber). Avoiding excessively large tubes also minimizes complications during use, but the tube must be long enough to extend well into the oropharynx. An estimate of appropriate length is the distance from the tip of the nose to the tragus of the ear.

The use of nasopharyngeal airways has reduced dramatically with the rise in popularity of the LMA. They retain an important role in patients in whom the oropharynx cannot be accessed and in sedated patients who would not tolerate an oropharyngeal airway.

Complications

The commonest complication is trauma causing bleeding, which can be extensive and may compromise the airway further. They should not be used where there is evidence of coagulopathy or other potential causes of epistaxis. Inappropriate sizes may either fail to clear the airway (too short) or lead to coughing, laryngospasm or retching (too long).

FACEMASKS

Anaesthetic facemasks are designed to fit over the patient's nose and mouth and enable the creation of a low pressure seal. This should not require excessive force. The facemask (Fig. 6.7) consists of three parts: the mount, the body and the edge. A snug fit is achieved by incorporating one or more of the following features into the design: by anatomically shaping the body, by the use of an air-filled cuff at the edge (Figs 6.8B, D and E) that has a soft cushioning effect or by a soft pliable flap (Figs 6.8A and C) that takes up the contour of the face.

The mount should be a 22 mm female taper if made to the standards of ISO (International Organization for Standardization) or BS (British Standards). It is usually

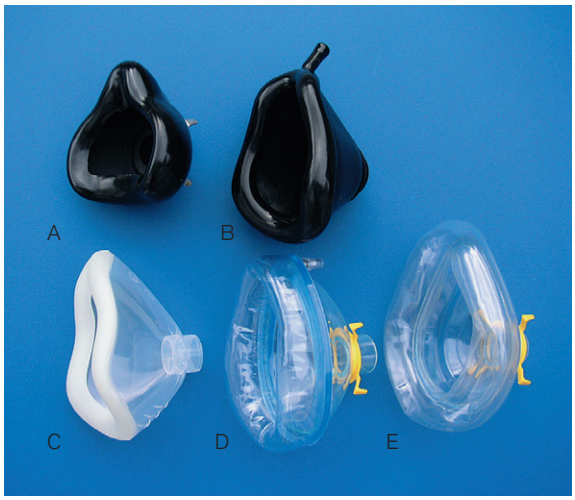


Figure 6.8 A selection of facemasks: **A.** black rubber 'Everseal' mask with soft pliable flap (designed by MIE Ltd in 1953 for the British Everest expedition); **B.** 'anatomical' facemask in black rubber with air filled cushion; **C.** single-use facemask with flap; **D.** single-use facemask with adjustable air filled cushion; **E.** single-use facemask with pre-filled air cushion. Note: **A**, **D** and **E** carry attachments for use with a harness (see text).

constructed of hard synthetic rubber but may be plastic or metal. The former two wear more easily with repeated use and eventually produce a leak or potential for accidental disconnection.

The body may be made from black rubber, neoprene, plastic, polycarbonate or silicone rubber. In some cases, a malleable wire stiffener or wire gauze is incorporated in the body so that the shape may be altered to fit the patient's face. The transparent body of a polycarbonate or plastic facemask permits continuous inspection of the airway and respiration to be monitored by the appearance of condensation during exhalation. This is useful during anaesthesia and particularly during resuscitation. A transparent facemask also affords the possibility of seeing regurgitant matter or vomitus emerging from the mouth. Such masks are perhaps less threatening to children and anxious adults (Figs 6.8C, D and E).

The internal volume (apparatus dead space) within the body of the facemask is relatively unimportant in adults but may assume significance in neonates and infants where it can constitute 30% or more of their tidal volume. Several designs shape the paediatric facemask to minimize the apparatus dead space (Fig. 6.9). In spite of this, good fit may be more important than theoretical apparatus dead space: paradoxically in paediatrics, a larger facemask particularly of the Rendell-Baker or Laerdal type, may allow the face to fit further into the mask with improved fit and reduced effective apparatus dead space.

The edge may be anatomically shaped and fitted with a cuff or flap. A good fit is essential to prevent dilution of



Figure 6.9 The full range of Rendell-Baker paediatric facemasks, sizes 00, 0, 1, 2, 3.

administered gasses by room air during spontaneous respiration and to allow positive pressure ventilation without gas leak. A variety of types and sizes of facemask must be available, since none will be a good fit for every face. Edentulous or bearded patients may be especially difficult. The former are best managed by leaving any dentures in place to prevent the cheeks from falling away from the mask and by using a smaller mask. Beards often prevent a good seal around the edge of the mask and a leak-free fit may sometimes be achieved with a bigger mask held firmly with two hands. Anaesthetists have often had to go to bizarre lengths to achieve a useful seal in heavily bearded patients (e.g. using a pierced defibrillator gel pad on the face[3] or even wrapping the entire head in clingfilm[4]). Reusable masks with a cuff have a small filling tube fitted with a plug to enable the degree of inflation to be regulated. The plug must be removed to allow the cuff to deflate if the mask is to be autoclaved.

Whereas some facemasks withstand the high temperatures of autoclaving, others do not. Since these are not easily distinguished, many adopt uniform policies of disinfection. Care must be taken if chemical disinfection is used as some chemicals, e.g. chloroxylenol (Dettol), are known to have been absorbed by the material of the facemask and have resulted in injury to the patient's skin.

The shift towards disposable single-use facemasks avoids the cost of sterilization (increasingly performed off-site for many hospitals) packaging, and mandatory tracking and trace systems, but risks a reduction in quality. The same principles of design nevertheless apply to plastic disposable facemasks (Figs 6.8C, D and E). Materials and components are cheaper and of lower quality, but more importantly there is a much more limited range of designs and sizes. Most, even those for paediatric use, are essentially based on the one design of air-filled cuff and cone-shaped body. As they are reproduced with differing quality and materials by numerous manufacturers, uniformity of performance cannot be assumed. They may be advertised with bold claims of high performance and popularity but formal clinical evaluation is not a pre-requisite to bringing them to market. Several new designs have been found to be substandard and it is wise to assess performance before changing devices. Sizes do not necessarily equate between manufacturers. Some single-use masks do not allow adjustment of the volume of the air-filled cuffs and these may be both under filled and of less pliable plastic leading to a poor-quality seal. Poor design or poor fit, even in the absence of a leak, can cause areas of high pressure on the skin, which if unchecked, could lead to ulceration. Single-use masks do, however, have a number of advantages beyond sterility. These include use of inert plastics (eliminating risk of allergy to latex), transparency and even the potential to add scent to the plastic making the products more readily accepted by patients.



Figure 6.10 A McKesson comfort cushion nasal inhaler.

Many anaesthesia masks are supplied with a ring device that has several lugs protruding to allow attachment of a head harness. Older black rubber masks have a similar metal harness ring (Figs 6.8A, D and E). A head harness may be used to allow facemask anaesthesia while allowing the anaesthetist to keep both hands free. Since the introduction of supraglottic airways head harnesses are rarely used, if ever, except during non-invasive ventilation (see below).

Masks for some dental anaesthetic techniques are designed to fit the nose only, so that the dentist has unimpeded access to the mouth (Fig. 6.10). They are also known as nasal inhalers.

are simple airway tubes (*first generation*), from those which are designed with the intention of reducing the risk of pulmonary aspiration of gastric contents (*second generation*). The proposed advantage of this classification is that it is simple, is not dependant on patient/device anatomy, but is 'patient centred'. Second-generation devices may be designed to improve patient safety, but the effectiveness of this design feature is not proven in all cases. Second-generation devices comprise the ProSeal LMA, the Combitube and Easytube, the Laryngeal Tube Suction mark II, the Streamlined Liner of the Pharynx Airway, the i-gel and the Supreme LMA.

Pharyngeal seal and efficacy vs oesophageal seal and safety

Functionality of an SAD depends on several factors including insertion ease and success rate, manipulations required during anaesthesia to maintain airway position and patient tolerance of the device during emergence. During controlled ventilation efficacy is dependent on factors such as whether the device orifice sits over the larynx and the quality of the device seal with the laryngopharynx (*pharyngeal seal*). The pharyngeal or airway seal pressure is usually assessed by allowing a fresh gas flow of 3–5 l min^{-1} into the closed breathing system of an apnoeic non-paralyzed patient and noting the maximal airway pressure generated or the pressure when a leak can be detected.

The safety of a SAD reflects the likelihood and severity of complications occurring at all stages of anaesthesia and afterwards. Complications include failures, displacements, airway obstruction and sequelae such as sore throat, pharyngeal trauma and nerve injuries from pressure effects. The risk of aspiration is a major concern with SADs. Moderation of this risk requires a good-quality seal with the hypopharynx and/or oesophagus (*oesophageal seal*) to prevent gas leaking into the oesophagus and stomach and also to prevent regurgitant matter passing from the oesophagus into the airway. Ideally oesophageal seal pressures are assessed in terms of hydrostatic pressure needed in the oesophagus to cause liquid regurgitation (hence needing cadaver studies). A correctly functioning drain tube should enable regurgitant matter to bypass the larynx and be vented externally. This protects the airway and gives an early indication of the presence of regurgitation.

Several recent studies indicate that the extent of oesophageal seal varies considerably between different SADs.[12–15] They also demonstrate that under experimental conditions in cadavers, those with a drain tube will usually effectively vent regurgitant fluid provided the drain tube is not occluded. Particulate matter has not been studied. These are important findings, though their clinical correlates are not in all cases confirmed.

First-generation SADs

The laryngeal mask airways

The next section focuses on the classic LMA. Much of what is included also applies to other laryngeal masks and supraglottic airways in general. Several alternatives to the classic LMA are described in more detail below.

The LMA Classic

As its name implies the classic laryngeal mask airway (cLMA) (Intavent Direct, Maidenhead, UK) is designed to act as a mask that fits over the larynx. It consists of an oval soft silicone mask that sits over the larynx with an integrated stem that extends through the oral cavity to allow attachment to the anaesthetic circuit or other appropriate equipment.

The mask (distal) end of the cLMA is made of medical grade silicone and consists of a shallow bowl resembling a small facemask, which is surrounded by an inflatable tubular cuff (Fig. 6.11). The latter, when inflated, fits around the laryngeal inlet and supports it in a position away from the posterior pharyngeal wall. The back of the bowl leads into a semi-flexible tube which passes out of the pharynx and mouth and has a 15 mm ISO male connector so that it can be attached to a breathing system. At the point the tube enters the mask, there are two thick silicone rubber strands (grilles, or bars) designed to prevent the epiglottis falling into it and occluding the lumen. The mask cuff is inflated via a pilot tube that terminates in a small 'pilot balloon' giving an indication of cuff inflation/deflation. A self-sealing valve prevents deflation of the cuff.

cLMAs are made from silicone rubber so that they can be autoclaved and reused. The manufacturer has designated the cLMA for 40 uses. Experience suggests

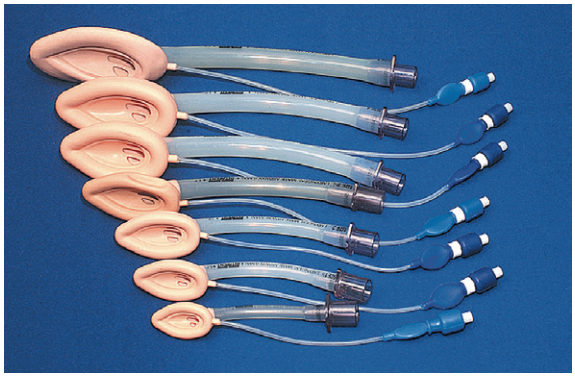


Figure 6.11 The laryngeal mask airway in sizes 1, 1½, 2, 2½, 3, 4, 5. A size 6 is now also available which is not shown here.

Table 6.1 Manufacturers recommendations for LMA selection and cuff inflation

LMA SIZE	PATIENT SIZE (kg)	MAXIMUM CUFF VOLUME (ml)
Size 1	Neonates up to 5	4
Size 1½	Infants 5–10	7
Size 2	Children 10–20	10
Size 2½	Children 20–30	14
Size 3	Patients 30–50	20
Size 4	Patients 50–70	30
Size 5	Patients 70–100	40
Size 6	Patients over 100	50

the cLMA can withstand many more cycles of sterilization, but this is 'off label' use and cannot be recommended. Originally produced in four sizes, two mid-range sizes and two further larger sizes were added later (Fig. 6.11).

Experience with sizes 1 and 6 is comparatively limited. It should be noted that, although originally developed from plaster casts of cadaveric adult larynxes, subsequent sizes are simply scaled versions of the originals. As infant and paediatric laryngeal anatomy varies considerably, the smallest sizes may not provide as reliable an airway as adult sizes. The manufacturer's recommendations are as shown in Table 6.1. Where the predicted size does not fit well, an alternative size may provide a better airway.

The term LMA is a registered trademark of Intavent Ltd, UK.

cLMA dimensions relevant for its use as a conduit

The size of tracheal tube that can be passed through a given size of cLMA varies as tracheal tubes of the same internal diameter can have different external diameters depending on the model and manufacturer (Table 6.2).

Table 6.2 Size of tracheal tubes that can be passed through a cLMA. The tabulated results are for a lubricated Portex blue line uncuffed tracheal tube* passed without force in a cLMA. If using a cuffed tracheal tube, reduce the size of tube by 0.5 mm. The results do not apply to laryngeal masks other than the LMA Classic. Distances are to the mask grilles: intubation would require longer tracheal tubes. (*External diameter ranges from 1.3 to 2.6 mm greater than internal diameter (ID), increasing as internal diameter rises.)

CLMA SIZE	SIZE OF UNCUFFED TT PASSABLE (mm ID)	DISTANCE FROM CONNECTOR TO MASK GRILLES (mm)
1	3.5	108
1½	4.5	135
2	5.0	140
2½	6.0	170
3	6.5	200
4	6.5	205
5	7.0	230

Inserting the LMA

Because the LMA, when correctly placed,would elicit a gag reflex in the awake patient, it should only be inserted in a patient whose pharyngeal reflexes have been sufficiently depressed by general anaesthesia or adequate local anaesthesia and/or analgesia.

The LMA is correctly placed when it is advanced to lie with its tip at the top of the oesophagus (surrounded by the upper oesophageal sphincter: cricopharyngeus); providing it is facing forwards the orifice of the mask will then lie over the laryngeal inlet.

Pre-use checks As the device is reusable and subject to wear and tear, before use it should be checked for damage, particularly cuff leaks, eccentric inflation of the cuff or failure of the self-sealing valve on the pilot balloon. The airway tube should also be checked by flexing to ensure it does not occlude when bent. Finally the lumen of the device must be checked to exclude the presence of foreign bodies introduced during the cleaning/sterilization process. Before insertion the concave part of the mask is pressed against a hard surface while deflating it, which causes the cuff to retract backwards behind the bowl. The back of the mask is lubricated with a water-based gel.

Standard insertion General anaesthesia should be adequate to allow generous mouth opening and jaw thrust without response. Under such circumstances there should be no need to use neuromuscular blockers. The patient's head and neck are then placed in the 'sniffing position'. The manufacturer's recommended insertion technique requires that the cuff is deflated as above and the LMA is grasped like a pen in the dominant hand. The tip of the operator's gloved index finger is placed at the junction of the tube and mask whilst the non-dominant hand maintains the position of the head and neck by cradling the occiput so that the patient's mouth falls open. The mask is inserted into the mouth and the bowl is kept pressed against the hard palate as it is advanced in one smooth movement into the hypopharynx. The upward pressure against the palate flattens the cuff of the mask to give a smooth thin leading edge. The hard and then the soft palate and finally posterior pharyngeal wall act as a scoop to guide the mask into place and prevent snagging on the tongue or epiglottis. The mask is advanced until resistance

is felt. For many people the index finger is not long enough to fully insert the LMA: the hand in the mouth, guiding the LMA, should remain in place while the hand on the occiput can now be used to push the LMA stem inwards (still guided by the hand in the mouth) until resistance is felt. The LMA is not fully inserted until it reaches resistance, signifying reaching the cricopharyngeal constrictor. Without holding the tube the cuff is then inflated with air. The manufacturer indicates a *maximum volume* for cuff inflation which must not be exceeded; inflation to an *optimal pressure* is preferable (see below). Use of the maximum volumes is likely to lead to excessively high intracuff pressures that reduce the device's efficacy and safety profile. Where inflation to a predetermined volume is used, half the manufacturer's recommended maximum is a good starting point.

From personal observation, most users do not use the recommended insertion technique and yet find that the device seats well and provides a reliable airway. It is perhaps this feature that accounts for the success of this device.

Alternative methods of insertion Many users simply grasp the airway near its proximal connector and slide the device down the back of the tongue, relying on the elasticity of the epiglottis to return it to its normal position and not remain down-folded over the larynx. Rotational techniques have been described in which the LMA is inserted either upside down and rotated 180° on reaching the soft palate (akin to insertion of a Guedel airway) or inserted laterally parallel to the tongue and rotated 90° inwards and towards the midline as the faucial pillars are reached. All alternative techniques are supported by some, but limited, clinical evidence, and may be particularly useful in children. Some techniques appear to be designed mainly to avoid insertion of the anaesthetist's hand into the patient's mouth.

It is reasonable to state that poor LMA insertion technique is common and that good basic technique improves anatomical placement and device function: this in turn is likely to improve device safety. Good basic technique is particularly important when the newer LMAs (flexible LMA, ProSeal LMA) and techniques such as controlled ventilation and tracheal access via the LMA are to be used.

Additional airway manoeuvres may be used to 'create pharyngeal space' while inserting the LMA. These include chin lift/jaw thrust (which can be applied by an assistant or by the operator by placing a thumb into the mouth and pulling on the jaw from behind the front teeth), or traction on the tongue. Any technique that requires the operator to remove their hand from the occiput risks losing the optimal head and neck position.

Confirmation of correct placement Remarkably, for a device that is inserted blindly, the mask almost always adopts the correct position and provides a patent airway with a success rate above 95%.[16]

When the LMA cuff is inflated three observations assist confirmation of correct placement. First as the mask tip inflates the LMA rises 0.5–2 cm before coming to an abrupt halt, second the anterior neck is seen to slightly fill and finally the longitudinal black line running along the dorsal aspect of the tube should remain in the anatomical midline. While none of these 'tests' are foolproof, any failures should raise suspicion that the mask is malpositioned. In particular, rotation of the longitudinal line generally indicates rotation or misplacement of the mask portion.

In a spontaneously breathing patient, ventilation should be silent. Airway noise (which may mimic stridor or bronchospasm) suggests misplacement with partial airway obstruction. Airway obstruction may arise from poor positioning or laryngospasm (often associated with an inadequate depth of anaesthesia). The anaesthesia reservoir bag excursion should be normal. Spirometry, available on many modern anaesthetic machines, is a useful monitor and shows a typical non-obstructed loop. In an apnoeic patient, gentle squeezing of the reservoir bag should produce normal chest movements with an applied pressure no greater than 20 cm H_2O. A small leak is permissible; a large leak or a high inflation pressure usually indicates the possibility of misplacement (frequently down-folding of the epiglottis or inadequate depth of insertion of the LMA), or of breath-holding by the patient.

The seal between the pharynx and the LMA is modest, with a median of 16–20 cm H_2O and rarely exceeds 30 cm H_2O. When controlled ventilation is applied to the LMA, airway pressure should not exceed 20 cm H_2O. Increasing airway pressures lead to loss of ventilating volume (risk of hypoventilation) and an increasing likelihood of oesophageal/gastric inflation (risk of regurgitation and aspiration). For this reason, the LMA is arguably not suited to use for controlled ventilation in obese patients and for those in challenging circumstances such as lithotomy position and for laparoscopic surgery. The main reason is that there are better and safer SADs available for such uses.

Indications for using the LMA:

Instead of a facemask and pharyngeal airway, with the added benefit of:

1. releasing the anaesthetist's hands to deal with other matters, and removing them from proximity to the operative site
2. providing a more reliable airway, particularly in patients placed in positions other than supine or where a facemask does not seal well against the face as in bearded patients.

Instead of a tracheal tube, in certain circumstances, for example:

1. elective procedures involving spontaneous or controlled ventilation, where tracheal intubation would have been used only to enable a 'hands-free' and gas-tight airway

2. elective procedures where intubation and the necessary drugs may elicit particularly undesirable cardiovascular or respiratory responses
3. patients in whom difficult intubation (but not difficult airway or mask ventilation) is predicted, as in those with cervical spine problems or moderately limited mouth opening, usually as a prelude to intubation via the LMA
4. as a first choice airway for use in emergencies by non-anaesthetic staff on the basis that success rate at establishing ventilation for non-specialist staff is greater with the LMA than with facemask ventilation or tracheal intubation.[8]

As an aid to extubation:

1. The LMA can be used as an aid to extubation and recovery, even after peroperative tracheal intubation. The LMA is less stimulating than a tracheal tube and less likely to elicit coughing and straining during recovery. It also protects the larynx from blood and secretions arising from the nasopharynx and oropharynx. An example of such use would be insertion of the LMA at the end of a tonsillectomy to facilitate a smooth wake-up. Placement of the LMA is much easier if the tracheal tube is at first left in place to guide insertion of the LMA. The device is advanced over the posterior aspect of the tracheal tube which ensures that the tongue and epiglottis are easily negotiated and the larynx may be held forward allowing the tip of the LMA to seat more easily behind the cricoid. The tracheal tube can then be removed.

For difficult airway management:

1. To provide a *conduit* to facilitate intubation. The LMA orifice lies directly over the laryngeal inlet in >90% of cases. In theory a gum elastic bougie or narrow long tracheal tube may be passed blindly through the LMA to enter the trachea; however, in practice, success rates are as low as 10–25% and such a 'blind technique' also risks trauma to a potentially already compromised airway. Far preferable are flexible endoscopic guided techniques, with use of an Aintree catheter being the best technique (described below).
2. To *rescue* the airway when intubation and/or facemask ventilation have failed. There is now considerable evidence to support the use of the LMA in these scenarios, some even suggesting that the LMA is more likely to be an effective airway in those in whom intubation has failed than it is in the general population. The device is consequently recommended in all international airway management algorithms.

During resuscitation:

1. There is considerable interest in use of SADs for maintenance of the airway during cardiopulmonary resuscitation. Several SADs are now recommended by the resuscitation councils. The largest bulk of evidence supports the use of the LMA, though several newer SADs that enable a higher pharyngeal seal may offer additional protection against regurgitation/aspiration and have potential benefits. The LMA may be used as a primary airway by those with little or no intubation skills and as a rescue device when tracheal intubation fails or is impractical because of patient entrapment. It may be easier for non-anaesthetists to achieve ventilation with a SAD than with a facemask.[17]

Contraindications

Accepted wisdom has it that the LMA does not protect the lungs against aspiration of refluxed or regurgitated gastric contents. In reality it offers some protection in the unconscious patient and therefore is 'safer' in this respect than facemask ventilation, with or without simple airway adjuncts.[13,15] However, the LMA does provide less protection than the tracheal tube. The LMA should, therefore, not be used in patients with a full stomach or risk factors for regurgitation/aspiration (delayed gastric emptying, hiatus hernia, etc.).

The LMA is also contraindicated in patients who are difficult to ventilate by virtue of size, chest disease or surgical positioning, as the mask seal is rarely above 25 cm H_2O.

The LMA is more susceptible to dislodgment than a tracheal tube and because it lies external to the larynx, laryngeal closure (e.g. laryngospasm) will lead to airway obstruction. The LMA should therefore not be used during procedures where access to the airway for repositioning is impractical. It is unwise to use the LMA for prone surgery, especially if the patient cannot be rapidly turned supine. It is widely used for ear, nose and throat surgery and intraoral procedures, such as tonsillectomy but this requires experience in lower risk situations, good communication and co-operation with the surgeon and a high level of surveillance by the anaesthetist to detect and rapidly respond to malpositions or disconnections. Devices other than the classic LMA are likely to be more appropriate: especially the flexible LMA.

LMA Flexible

The reinforced (or flexible) LMA (Intavent Direct, Maidenhead, UK) is an alternative version of the LMA in which the tube is thinner, narrower and longer and is reinforced with a spiral of steel wire to add flexibility and reduce the risk of kinking; it is available in sizes 2–5 (Fig. 6.12). Placement of the flexible LMA (fLMA) requires meticulous technique to prevent rotation leading to mask misplacement: the mask can rotate 180° about the axis of the airway tube, leaving the mask orifice facing backwards, without this being evident proximally. In performance terms the main difference between the fLMA and the classic LMA is that, once placed, its proximal tube can be moved in any direction without those movements being

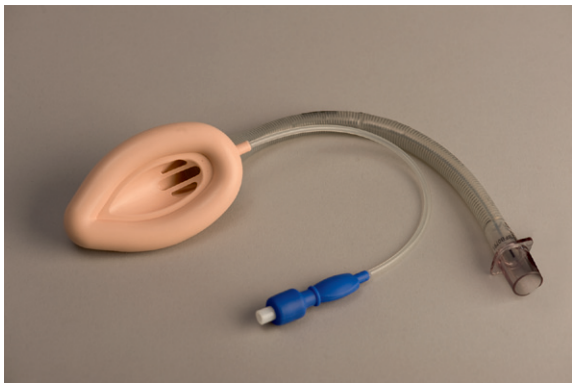


Figure 6.12 LMA Flexible, (Intavent Direct, Midenhead, UK).



Figure 6.13 A single-use laryngeal mask incorporating Cuff Pilot cuff pressure monitoring device.
Photograph courtesy of Ultimate Medical Pty, Victoria, Australia.

transmitted to the mask end of the device and leading to displacement. This makes the fLMA very popular for head and neck and intraoral procedures and enables the tube and breathing circuit to be positioned away from the operative site. Because the mask portion of the fLMA isolates the lower airway from above it is suitable for nasal and dental procedures. It may be used without a throat pack (therefore bypassing the risk of throat pack retention) though careful attention is required of both surgeon and anaesthetist to ensure surgical debris is not in the oral cavity during emergence. The fLMA is routinely used by a minority of anaesthetists for tonsillectomy where it can reduce the risk of blood entering the trachea during recovery from anaesthesia.

Because of the flexibility of the airway tube, insertion of the fLMA is more difficult. The bowl tends to rotate or the stem buckles if it is simply pushed from above as many do with the classic LMA. Alternative techniques of insertion specific to the fLMA have been described. Most involve use of a rigid or semi-rigid accessory (either placed within the fLMA tube, or attached to the outside of it) to increase the longitudinal rigidity of the device. Many, however, fail to prevent the commonest cause of misplacement of the device which is axial rotation. There is no convincing evidence that any of these techniques have benefit over standard insertion techniques and several introduce the possibility of trauma caused by the 'stiffener'.

Single-use versions of both the classic and flexible LMA are available. While these are identical in form many are made of PVC and therefore less pliable. Minor performance differences exist between reusable and single-use devices with the reusable devices generally performing better with easier insertion and less airway trauma.

Other 'laryngeal masks'

The term LMA is a registered trademark and applicable only to those devices manufactured by the company who originally produced the classic LMA. When the patent on many (but not all) aspects of the LMA design lapsed in 2003 several manufacturers introduced designs that are similar, though not identical (for example none had the epiglottic grill which remained under patent until 2008). Since that time there has been an explosion of such devices with more than 20 companies now distributing these products. To distinguish such a device from the LMA it should correctly be referred to as a *laryngeal mask* (LM). The majority of these devices will differ from the original LMA in some aspect. Mask shapes and sizes vary, as do the profiles of the airway tube (stem) and its connection to the bowl. They may be made of silicone, PVC or other plastics. Some devices have been introduced, withdrawn, redesigned and reintroduced. There has been much marketing but little evaluation of most new LMs. Design and material changes will inevitably alter function somewhat. In many cases this may not be clinically important. Whether these design changes alter the patient experience is largely unevaluated. Whether minor performance differences are important, when LMs are used for indications such as for airway rescue and for difficult airway management, is also generally unknown. What can be said is that most silicone devices have better performance characteristics than PVC ones and several designs do not perform as well as the original LMA design, with none, to date, exceeding its performance. In summary: it must not be assumed that generic and novel laryngeal masks will reproduce experience with the reusable LMA Classic or that they are fully interchangeable.

Cuff pressure monitoring

One novel design feature is an integrated cuff pressure monitor, currently unique to one manufacturer. The 'Cuff Pilot' (Ultimate Medical, Victoria, Australia) is a simple pressure indicating device incorporated into the inflation valve to give a constant visual representation of intracuff pressure (Fig. 6.13).

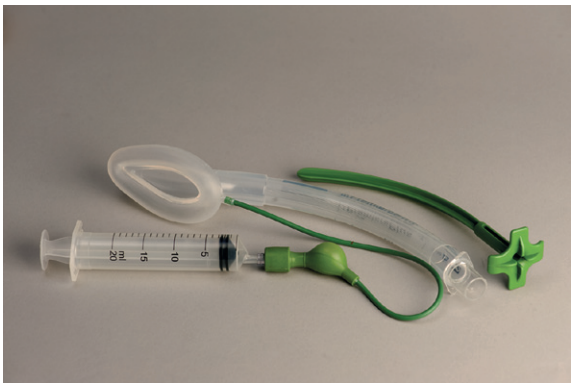


Figure 6.14 A single-use laryngeal mask with herniating pilot balloon to indicate an over-inflated cuff. A positioning aid is also shown which is inserted as a stiffener into the stem of the device. Armstrong Medical, Coleraine, UK.

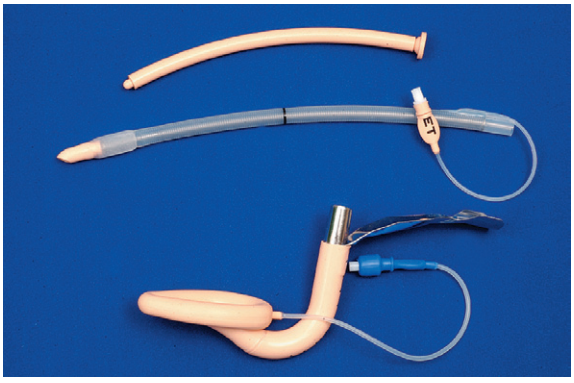


Figure 6.15 The intubating laryngeal mask airway (Intavent Ltd, UK). The LMA can be used with dedicated tracheal tube and 'pusher'.

One other manufacturer (Armstrong Medical, Coleraine, UK) produces a pilot balloon that simply and visibly herniates above recommended pressures (Fig. 6.14).

Intubating LMA

The intubating LMA (ILMA, Intavent Direct, Maidenhead, UK) was designed specifically to enable (blind) tracheal intubation by capitalizing and improving on the LMA's high success rate at guiding instruments within its lumen into the larynx (Fig. 6.15).[18,19] It is significantly different from the classic LMA: the bowl of the device is rigid as is the short, wide right-angled (approximately 110°) airway tube, which is constructed of stainless steel. This necessitates a novel insertion technique and a handle is provided to allow manipulation of the position of the bowl. Instead of grills across the mask orifice there is a semi-rigid flap designed to lift the epiglottis as the tracheal tube enters the bowl of the mask and prevent impaction of the tracheal tube. The device is supplied with a dedicated tracheal tube made with a soft rounded silicone 'bullet-shaped' tip to better negotiate the curve of the device and which is less traumatic when impacting the larynx. The ILMA tracheal tubes range in size from 6.0 to 8.0 mm ID, with all size tubes passing through all ILMAs.

Intubation through the ILMA involves four steps:

1. ILMA insertion
2. reattachment of the breathing circuit and manual ventilation to determine the optimal position (highest compliance, lowest resistance). ILMA position is adjusted as necessary
3. while maintaining the ILMA in the optimal position the ILMA tracheal tube is advanced into the trachea and correct placement confirmed
4. removal of the ILMA over the tracheal tube and reattachment of the tracheal tube to the breathing circuit.

After intubation it is recommended to remove the ILMA for all but the shortest procedures, as the rigid airway tube exerts high pressures on the surrounding mucosa (sore throat and hoarseness though usually mild are more frequent after use of the ILMA than the cLMA). Removal of the ILMA is achieved by using a 'tube stabilizer' (or the hub of a 5 ml syringe) to maintain the tracheal tube position, while the ILMA is withdrawn over it.

Although the ILMA lies over the glottis rather less frequently than other LMAs, successful blind intubation is enhanced by features that ensure the tracheal tube exits the mask at the correct angle and in the midline. With good technique first time success rate of 75% is achieved; increasing to 88–95% with two or more attempts.[20,21]

The ILMA should not be used for those patients with contraindications to laryngeal mask use. Its use should be avoided in patients with oesophageal disease, as blind intubation attempts can result in oesophageal intubation with a risk of injury and perforation.

Particularly with the increased availability of flexible endoscopes, the exact role for the ILMA is questionable, given that it cannot be relied upon as a blind technique in the difficult intubation scenario and that for use as a conduit for a flexible endoscope the view may be better through an ordinary LMA.[22,23] Experience with the device as an airway is limited and it cannot be recommended for routine use;[24] this, together with the different insertion technique compared to the classic LMA, compounds problems associated with its use.

Recently a single-use ILMA made of clear plastic and PVC has been introduced by the original manufacturers. The design mimics that of the ILMA and early studies suggest similar if not slightly poorer performance to the reusable device.

Other first-generation SADs

Even excluding single-use versions of devices there are currently more than 10 different varieties of supraglottic

airway available in the UK. Several others have been withdrawn from the market but it is likely that even more are under development. This situation reflects the considerable success of the cLMA and other SADs. Not all currently available SADs have clearly demarcated applications and a full analysis of their merits and demerits is beyond the remit of this chapter. Many of the issues touched upon in the section on the LMA are equally pertinent to these other products. The most important of these devices are briefly discussed here.

Laryngeal tube

The Laryngeal Tube (LT) (VBM Medizintechnik GmbH, Salz, Germany) was first made in 1999 and consists of a slim airway tube with a small balloon cuff attached at the tip (distal cuff) and a larger asymmetric balloon cuff at the middle part of the tube (proximal cuff). Both cuffs are inflated through a single pilot tube. After insertion it lies along the length of the tongue with proximal and distal cuffs lying in the oropharynx and oesophageal inlet, respectively (Fig. 6.16). Inflation creates a seal and ventilation occurs through orifices between the cuffs. The device is made of silicone and is reusable up to 50 times. Seven sizes are available for all patient ages and sizes, with colour-coded connectors. Size selection for sizes 0–2 is based on patient weight and for sizes 2.5–5 on height. For adults a size 3 is suitable for patients below 155 cm height, size 4 for patients between 155 cm and 180 cm tall, and a size 5 in patients 181 cm or taller. Adult sizes are supplied with a reusable silicone bite-block.

The LT has been modified several times with changes to the tip, balloons, pilot tubes and ventilation orifices.

The device is inserted with the head extended or in the neutral position and the cuffs deflated and lubricated. The tip is placed against the hard palate and advanced in the centre of the mouth until resistance is felt. If no resistance is felt depth indicators on the device indicate optimal and maximal insertion. The cuffs are over-inflated to a pressure of 80 cm H_2O and then deflated to 60–70 cm H_2O. Alternatively, large syringes are provided with suitable volumes for each size of LT indicated on the side of the syringe in colours corresponding to the colour codes of the LT (60–90 ml for sizes 3–5). Gentle hand ventilation is used to confirm a functional position. If ventilation is not ideal the LT may be advanced or withdrawn slightly until ventilation is optimum.

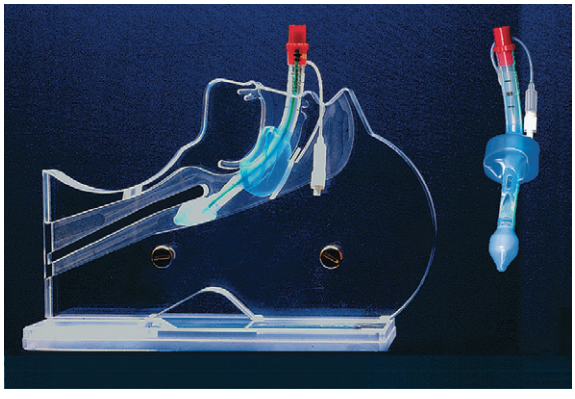


Figure 6.16 The Laryngeal Tube, VBM Medizintechnik, Germany.

The LT is designed for use during spontaneous breathing, controlled ventilation and for airway rescue.

At the end of anaesthesia the LT is left in place until the patient regains consciousness. The balloons are deflated as the device is removed.

The literature on the LT is somewhat confused by the multiplicity of versions that have been evaluated.[25] The slim profile of the LT allows easy insertion with little mouth opening and with the redesigned softer distal tip the device causes minimal airway trauma. Initial evaluations, with the original device, reported variable success rates: in particular spontaneous ventilation success was poor. More recent studies suggest high success rates for insertion (above 95%) and good airway seal pressure (approx. 26 cm H_2O).[9] It has a steep learning curve and may be suitable for use by non-anaesthetists and out of hospital.

The main limitation of the LT is that its tube-like shape provides little protection against axial rotation and this, combined with small ventilation orifices, tends to lead to a significant incidence of partial or complete airway obstruction. Compared to LMAs the LT is more likely to lead to airway obstruction, and less likely to lie over the larynx.[26]

Whether the lower cuff protects against aspiration of regurgitated stomach contents has not been determined, but a cadaver study of oesophageal seal demonstrated a high pressure seal, in the range 70–80 cm H_2O.

The poor view of the larynx, the narrow internal lumen and small airway orifices all make the LT rather poorly suited for tracheal access and exchange techniques compared to available alternatives.

A single-use version of the LT (LT-D) in PVC as opposed to silicone is available but has not been extensively evaluated.

CobraPLA

The Cobra Perilaryngeal airway (CobraPLA) (Engineered Medical Systems, Indianapolis, USA) is a relatively new single-use SAD (Fig. 6.17). The distal end (whose shape lends the name to the device) consists of a soft plastic head designed to seat in and seal the hypopharynx, with the anterior surface lying at the laryngeal inlet. The anterior surface of the head consists of a grille of bars soft enough to allow instrumentation of the larynx, but their presence prevents epiglottic obstruction of the device orifice. A proximal inflatable balloon is designed to elevate the tongue base and seal the oropharynx.



Figure 6.17 CobraPLA (Engineered Medical Systems, Indianapolis, USA).

The CobraPLA is supplied in eight sizes designed for use in patients ranging from neonate (size 1) to >140 kg (size 6). The internal diameter of the CobraPLA airway is larger than the corresponding cLMA: for the largest CobraPLA this is 12.5 mm.

Prior to insertion the cuff is deflated along the posterior aspect of the airway tube. The device is inserted until resistance is met as the distal tip reaches cricopharyngeus and it may then be withdrawn 0.5–1 cm. The proximal cuff is inflated until an airtight seal is obtained. The device is removed on return of consciousness with the cuff deflated.

The CobraPLA is designed for spontaneous and controlled ventilation and was modified in 2006. There is limited published experience and none with the largest and smallest sizes. Available evidence indicates 80% first attempt insertion success, airway seal of approximately 25 cm H_2O and minor morbidity of similar frequency LMAs.[27,28] A significant number of cases may require airway manipulation (pull/push) to maintain the airway.

As the CobraPLA is designed to be inserted to full depth and then withdrawn 0.5–1 cm, the tip likely does not obturate the oesophagus, which may be of significance in positive pressure ventilation: raised pharyngeal pressure being fully transmitted against the upper oesophageal sphincter. The pharyngeal cuff may prevent egress of regurgitant matter from the oropharynx, and hence increase the risk of aspiration.[29,30] With the exception of a modest increase in airway seal, there do not appear to be any benefits for use over the cLMA.

Second-generation SADs

Second-generation SADs have been designed to be safer for patients by use of additional features to minimize the risk of regurgitation and or aspiration.

LMA ProSeal

The LMA Pro-Seal (Intavent Direct, Maidenhead, UK) (PLMA), which was introduced in 2000, is an important addition to the family of LMAs.[31] It was designed with three refinements in mind: (1) improved performance during controlled ventilation, (2) improved safety regarding aspiration, and (3) an ability to diagnose misplacement of the device tip.

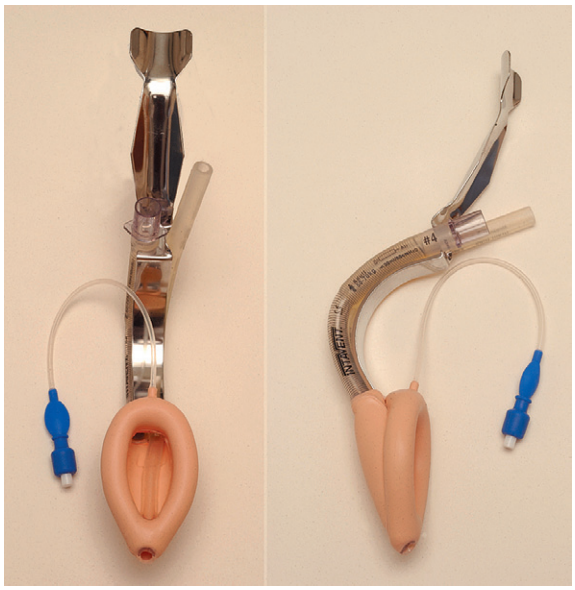


Figure 6.18 The LMA Pro-Seal with introducer attached.

The PLMA has a softer, larger and deeper mask bowl than the cLMA (Fig. 6.18). The mask cuff extends over the posterior aspect of the bowl pushing it forward when inflated. A drainage tube passes from the tip of the mask, through the bowl, to run parallel to the airway tube. Proximally these are joined by an integral bite block. The changes to mask shape, size and the posterior cuff increase the median airway seal by >80% to above 30 cm H_2O.[32] A full range of sizes is available with paediatric ones lacking the posterior mask cuff.[33]

The PLMA may be inserted digitally (like the cLMA), or by attaching a metal introducer (Fig. 6.14) or by railroading the whole device via its drain tube over a bougie placed in the oesophagus.[34] The latter technique is the most successful and least likely to lead to misplacement.[35] It is the technique of choice when first-time insertion success is critical, such as when managing a difficult airway. A gastric tube can be passed through the drain tube when indicated: this should be well lubricated and not refrigerated. The drainage tube allows such reliable insertion of an orogastric tube that inability to do so indicates the PLMA is misplaced with the posterior of the mask folded backwards. Although maximum volumes for cuff inflation are published by the manufacturers, inflation to an intracuff pressure of 60–70 cm H_2O is preferable.

A series of post-insertion tests are designed to confirm correct device positioning.[32] First a small amount of gel

occluding the proximal drain tube should not be displaced when a pressure of 20 cm H_2O is applied to the airway; this tests separation of the gastrointestinal and respiratory tracts and fails when the PLMA is not pushed in far enough, allowing gas to pass directly from the airway tube up the drain tube. Second, no more than one-third of the bite block should be visible as this also suggests incomplete insertion. Third, pressing on the chest should not displace the drain tube gel; if it does it suggests the drain tube tip has entered the glottis, though airway obstruction is likely to coexist. Finally, pressure at the suprasternal notch should lead to the drain tube gel bulging outwards; this tests that the drain tube is not folded over as suprasternal notch pressure is transmitted to the oesophagus and then to the drain tube, unless the tip is folded over. If doubt exists, attempting to pass a gastric tube (approximately 30 cm in adults) to the tip of the drain tube, or beyond will identify whether it is folded over. In practice these several tests can be carried out in a matter of seconds to confirm correct positioning and function of the PLMA with further attention only necessary if a test is not 'passed'.

Compared to other LMAs and LMs the PLMA creates a higher seal with pharynx and oesophagus. It enables more reliable controlled ventilation and increases the patients and indications where a supraglottic airway is appropriate. The PLMA also exerts less pressure against the mucosa than either cLMA or ILMA, so reducing the potential for mucosal trauma and damage. These advantages must be balanced against slightly greater difficulty in insertion of the PLMA. When combining a large number of studies, first-time insertion success with the PLMA is 85% and with the cLMA is 93%.[32,36] In contrast a bougie-guided technique is almost 100% successful. The PLMA has potential roles in difficult airway management and is perhaps the logical choice for rescuing the airway after failed rapid sequence induction.[35,37]

There is good theoretical and performance evidence to support the view that compared to the cLMA the PLMA does reduce gastric inflation and increases protection from regurgitated gastric contents. However, this is entirely dependent on correct positioning of the device. The PLMA lies with the drain tube in continuity with the oesophagus and the airway tube in continuity with the trachea. As the pharyngeal and oesophageal seals are both increased this creates functional separation of the gastrointestinal and respiratory tracts. The drain tube vents gas leaking from the mask towards the oesophagus and prevents gastric distension. It also vents regurgitated stomach contents. Design and performance features of the PLMA make it likely that its use reduces the risk of gastric inflation and protects from aspiration if regurgitation occurs. Extensive evidence from bench work, cadaver and clinical studies strongly support this claim, but it is unproven and probably unprovable in clinical use. Cadaver studies show an oesophageal seal of 70–80 cm H_2O and efficacy of the drain tube in venting regurgitated fluid.[14] There are several case reports of regurgitated matter being vented by the drain tube.

There are several studies comparing PLMA performance with other SADs. In all of these to date the PLMA performs as well as or better than alternative devices. Infrequent complications associated with the PLMA include oesophageal breathing, gastric insufflation and airway obstruction.[32]

In summary, the PLMA undoubtedly has improved performance characteristics over standard laryngeal masks with these being most marked when controlled ventilation is used. Whether it is safer than standard LMs and the classic LMA is impossible to prove in absolute terms but the evidence supports that view. Considering all the available evidence, the PLMA lessens the risk of aspiration compared to the classic LMA, and there is an argument for its routine use. However, the PLMA must not be used electively for cases of significant aspiration risk. Indications include 'extended roles' such as obese patients, laparoscopic surgery and selected open abdominal surgery. Use in such cases mandates good understanding of the device and its limitations, experience in lower risk cases and excellent technique. The limit of applications for the PLMA is not known but like any SAD the benefits and risks of its use, compared to a tracheal tube, should always be carefully considered before selection.

LMA Supreme

The LMA Supreme (Intavent Direct, Maidenhead, UK), the newest of the LMAs introduced in 2009, is a single-use SAD (Fig. 6.19) made of PVC and designed to combine the most useful elements of the PLMA (improved seal, drain tube, integral bite block) with the ease of insertion of the ILMA in a single-use device. The manufacturer's indications for use are the same as the PLMA. Despite this, it is not simply a 'a single-use' version of the PLMA as there are several important differences: the stem and mask are



Figure 6.19 The single-use LMA Supreme (Intavent Direct, Maidenhead, UK).

more rigid as is the bite block; the drain tube runs within the airway and divides it into two narrow channels and the bowl of the mask includes patented 'fins' that are designed to prevent airway occlusion by the epiglottis.

The SLMA is inserted with the patient in the 'semi-sniffing position' and is otherwise the same as ILMA insertion. After insertion, correct depth of insertion (and sizing) is indicated by a tab on the upper surface of the device, which should lie 0.5–2.0 cm from the upper lip. If it abuts the lip a larger size should be chosen. The tab is also designed for fixation of the device using adhesive tape.

Due to its newness the SLMA has been less completely clinically evaluated than other LMAs.[38] It is very easily and reliably inserted, even by novices.[17] Once inserted ventilation is reliable and the pharyngeal seal (approximately 24–28 cm H_2O) is higher than the cLMA but lower than the PLMA. The airway sits over the larynx as frequently as the cLMA or PLMA. An orogastric tube can usually be passed via the drain tube. In a direct comparison between the SLMA and i-gel (see later), performance of the two devices during simulated airway rescue was equivalent.[39] While efficacy in low-risk patients is therefore largely established, its further performance is not. The effectiveness of the drain tube is yet to be reported. Further evaluation is awaited before it can be recommended for the same extensive a range of indications as the PLMA but early impressions suggest the SLMA will have indications both in and out of hospitals for a wide range of indications.

i-gel

The i-gel (Intersurgical, Wokingham, UK) is a novel single-use, cuffless SAD made of a medical-grade elastomer gel (styrene ethylene butadiene styrene). Its shape partially resembles the inflated PLMA. It has a short wide-bore airway tube with no grilles, an elliptical shaped stem, an 'anatomically' shaped bowl, an integral bite block and a drain tube (Fig. 6.20). These features provide low resistance to gas flow, stability, improved pharyngeal seal, the potential for good access to the airway as a conduit, and possibly decreased risks of airway occlusion or aspiration.

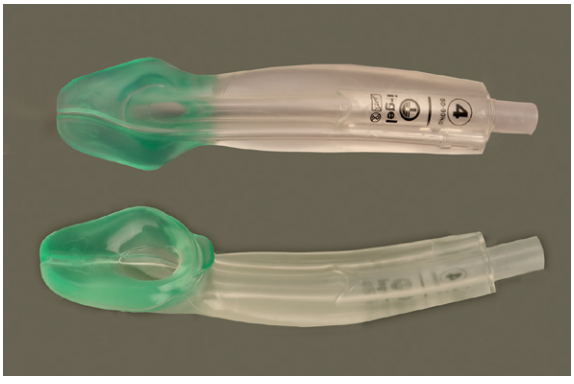


Figure 6.20 i-gel disposable supraglottic airway (Intersurgical, Wokingham, UK).

The i-gel is inserted in the sniffing position and after lubrication of the back, front and sides of the device. Standard insertion mimics cLMA insertion with the passage of the i-gel following the roof of the mouth and posterior pharynx until stopped by cricopharyngeus muscles. A rotational insertion technique is also described.

The i-gel is very easily inserted, by experienced and novice users, due to both low frictional properties and the lack of a cuff needing inflation.[40,41] The pharyngeal seal is 24–28 cm H_2O in most cases, but in a minority is considerably lower and ventilation is not possible. An alternative size should be tried before abandoning the device. Several authors describe that the elastomer's 'thermoplastic' properties leading to an improving seal over time as the device warms, but this is an inconsistent finding and it is also seen with cuffed devices. It is just as likely due to a combination of adaption of the pharynx to the shape of the device within it, and better relaxation with deeper anaesthesia.

The i-gel offers the potential for improved ease of use, improved ventilation and increased safety compared to the cLMA, in a disposable SAD. Its design and performance characteristics make it an alternative to the cLMA. Cases of airway rescue and use of the i-gel to facilitate fibreoptic-guided intubation are reported.

Although the drain tube is smaller gastric access is as reliable as for the PLMA, but the i-gel has lower pharyngeal (ventilation) and oesophageal (protection from regurgitation) seal pressures than the PLMA.[14,38] Whether this has any clinical importance in practice is not yet clear. The same studies report the drain tube is effective in venting regurgitant fluid, unless it is blocked.[14,42] The i-gel was intentionally designed with a truncated tip to reduce compression of the oesophageal sphincter and so reduce dysphagia, which may account for the reduced oesophageal seal. On the available evidence it would seem prudent to remove any orogastric tube after use as the i-gel oesophageal seal may not protect the airway if the drain were to become blocked. Cases of protection from aspiration with the i-gel are reported, as is one case of partial aspiration.[43]

Complications reported with the i-gel are few. Laryngopharyngeal trauma and pain after use appear infrequent indeed. Transient nerve injuries and lingual congestion have been reported rarely.

Combitube

The Combitube (Covidien, Ireland) originally marketed as the Oesophageal Tracheal Combitube (Fig. 6.21) is a development of the oesophageal obturator airway. It is a single-use device and is conceptually different from other

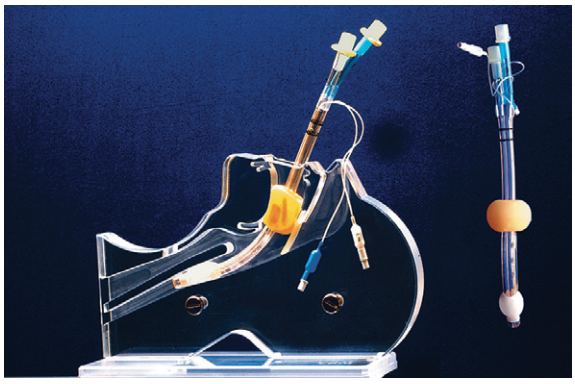


Figure 6.21 The Oesophageal Tracheal Combitude, Kendall, USA.

SADs in that it is designed to provide an airway after blind placement in either the trachea or the oesophagus. It has two cuffs (like the Laryngeal Tube) and two tubes. The lumen of one tube opens beyond the distal cuff, whilst the other ends between the two cuffs and has only side openings. There were originally two sizes, 37 Fr and 41 Fr, for small and larger adults. A smaller size 26 Fr was recently introduced along with other modifications aimed at reducing the incidence of trauma.

With the head and neck *flexed* or in the neutral position it is inserted blindly into the mouth and advanced until the teeth reach between the depth marks. First the proximal high-volume pharyngeal cuff is inflated with 90–100 ml of air to fix the device in place then the lower cuff is inflated with about 15 ml. When inserted blindly, the Combitube almost invariably enters the oesophagus. The proximally longer tube, marked '1', has the lumen ending between the cuffs and ventilation is first attempted through this lumen. If this fails then the second lumen is tried. If oesophageal insertion is confirmed, tube '2' can be used to insert a gastric drain with the smaller balloon acting as an oesophageal obturator. An oesophageal detector device, such as Wee's device, can be used to confirm the position of the tube tip before ventilation, thereby preventing inflation of the stomach in those cases where the device enters the trachea. Where ventilation with the Combitube fails, this may be due to obstruction of the larynx by the oropharyngeal balloon; the device should be withdrawn by 1–2 cm and ventilation attempted again.[25] The incidence of trauma with the Combitube is higher than other SADs and oesophageal rupture must be considered an inherent risk;[44,45] the mechanism may involve raised oesophageal intraluminal pressure as a result of obturation, as well as direct injury by the device.

If tracheal intubation is attempted at any time after insertion, the proximal Combitube cuff is deflated, the tube pushed to the left-hand side of the mouth and conventional laryngoscopy performed. The oesophageal seal provided by the distal cuff should prevent soiling of the pharynx by gastric contents during the procedure.

Due to its cost, trauma associated with insertion and the potential confusion of the dual tubes, the Combitube has been superseded by other SADs and has no role in routine elective anaesthesia. It is targeted primarily now at emergency airway management and non-anaesthetic use. The Combitube may have a role in out-of-hospital airway rescue in those trained in its use. It is part of the ASA difficult airway management algorithm and ILCOR recommendations for airway control during cardiopulmonary resuscitation.[46]

The *Rusch Easytube* (Teleflex Medical GmbH, Kernen, Germany) is a very similar, but more recently introduced device. Minor modifications, including a less bulky distal tube, suggest levels of trauma associated with its use may be reduced, but there is, to date, limited information.

The Laryngeal Tube Suction mark II

In 2002 the Laryngeal Tube - Suction (LTS) (VBM Medizintechnik, Salz, Germany) was introduced as a modification of the Laryngeal Tube, with a drain tube lying posterior to the airway tube to enable gastric tube placement and prevent gastric inflation during ventilation. While the design aimed to increase device safety (*compare* cLMA and PLMA) the new device was much more bulky, harder to introduce, and potentially traumatic: thereby losing two of the LT's principal advantages.[47] In 2005 the mark II LTS (LTS II) was introduced with a slim profile tip and an asymmetric oesophageal balloon. Size selection, insertion technique and reusability for the LTS-II are identical to the LT.

The LTS-II offers the potential for expansion of the role of the LT. It is easier to insert than the LTS and its pharyngeal seal is similar to the PLMA. Evidence of ease of insertion and incidence of airway obstruction remains inconsistent, but is generally improved.[48]

Like the LT, in cadavers, the LTS-II seals well with the oesophagus to a pressure of 70–80 cm H_2O and the drain tube effectively vents regurgitant fluid.[15]

Considerably more evidence is needed before the LTS-II role is fully established.

A single-use version of the LTS-II exists (confusingly) called the LTS-D.

SLIPA

The Streamlined Liner of the Pharynx Airway (SLIPA) (SLIPA Medical, Isle of Man, UK) is an inexpensive single-use SAD of novel design, a blow-moulded airway made of a polyethylene based composite with the shape mimicking a 'pressurized pharynx' (Fig. 6.22). It was first launched around 2003, although issues with choice of manufacturer and distributor continue to affect availability in the UK and independent evaluations of the device are so far

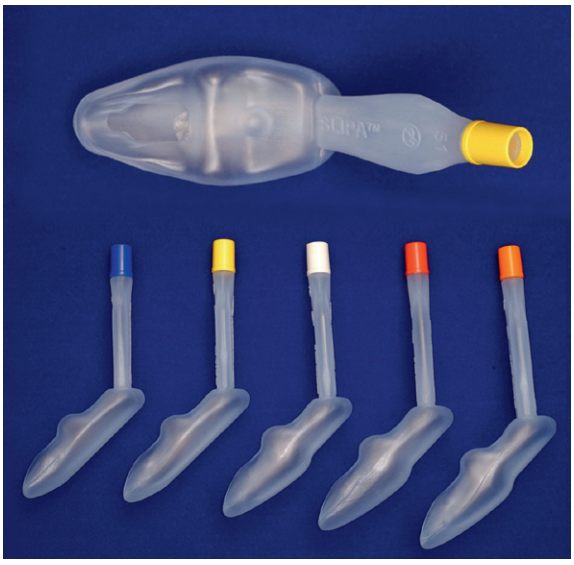


Figure 6.22 SLIPA (SLIPA Medical, Isle of Man, UK) disposable supraglottic airway device in a range of adult sizes.

scarce. The SLIPA slightly resembles a boot. It has no cuff, is hollow and is designed to sit with the toe of the boot in the hypopharynx. Lateral prominences in the mid-portion of the airway are designed to locate in the pyriform fossae and the mask displaces the tongue base anteriorly, with the intention of producing a stable device with an improved airway seal.

The SLIPA is included in the second-generation SADs on the basis of design only. The inventor reports that the interior of the SLIPA provides a protective reservoir able to accommodate 50–70 ml of fluid (compared to a 5 ml capacity of the cLMA).[49] There are no clinical data to support the clinical relevance of this claim.

There are six adult sizes (47, 49, 51, 53, 55 and 57) and while size selection is based on height, the aim is to match the maximum diameter of the SLIPA with the maximum diameter of the larynx (measured as the maximum width of the thyroid cartilage in millimetres).

Insertion of the SLIPA is aided by extension of the head and neck and may be improved by lifting the mandible or by use of a laryngoscope or finger placed in the pharynx to increase pharyngeal space. Repeated attempts may be associated with increased minor airway trauma.

Early cohort and comparative trials by the inventor showed satisfactory insertion performance.[50] A few small independent trials have been performed which are supportive of its use showing adequate insertion success and pharyngeal seal pressures similar to the LMA.[51,52] Trauma and bleeding are potential concerns.

There are no data available about oesophageal seal and no clinical trials addressing airway protection.

SADs SUMMARY

The topic of SADs is complex with too many devices and variants of each. The LMA Classic is a highly reliable and safe device and remains the SAD against which all others should be benchmarked.[53] For many newer SADs there is too little research to perform benchmarking and where appropriate evaluations have been performed several first-generation SADs (airway tubes) appear to offer little, if any, benefit over the classic LMA.

In contrast the second-generation SADs do offer potential benefits, as they are all designed to increase safety over and above that offered by the SAD. Most have a higher pharyngeal seal than the classic LMA. Most also have drain tubes whose efficacy has been demonstrated. Several have increased oesophageal seal compared to the classic LMA, which raises the possibility, but not certainty, of increased safety.

TRACHEAL TUBES

It is often stated that 'the gold standard' for securing the airway is tracheal intubation. Improvements in SAD design and technology and increased recognition of the complications of laryngoscopy and tracheal intubation have given pause for thought on the veracity of this statement. However, it remains true that a tracheal tube, once placed, offers both reliability and protection that few, if any, SADs can currently provide.

A cuffed tracheal tube, once placed, provides the greatest likelihood of protection for the lungs from inhalation of foreign material, the tube is the least likely device to be dislodged, glottic reflexes are bypassed and for intra-oral and head and neck surgery, only a tracheostomy can potentially give better surgical access (see section above entitled 'Indications for using the LMA', for comparison).

History

Tracheal intubation in the form of tracheostomy pre-dates anaesthesia by a number of centuries.[54] As early as 1542, Vesalius[55] recorded intermittently blowing into a reed that was passed into the *aspera arteria* of an animal whose thorax had been opened. He found that this caused the lungs to expand, and the heart to recover its normal pulsations. In 1667, Robert Hooke, at the Royal Society in London, similarly kept a dog alive for over an hour by ventilating its lungs with a pair of bellows tied into the trachea which had been severed below the epiglottis. Thereafter artificial respiration by intubation of the trachea became fairly common by the end of the eighteenth century, for treatment of asphyxia and drowning.

A number of milestones are notable. In 1871, Friedrich Trendelenburg[56] developed a cuffed catheter for insertion through a tracheostomy to prevent soiling of the lungs during operations on the upper airway. This tube with its inflatable rubber cuff would look familiar to any modern day anaesthetist and was widely used for the next 30 years.[57] In 1878, the Glasgow surgeon William MacEwen[58] placed a metal tube by manual palpation through the mouth into the trachea of a 55-year-old plasterer and, following the administration of chloroform, packed off the laryngeal opening to successfully resect a tumour at the base of the patient's tongue. 'Endotracheal anaesthesia' was born. Eisenmenger's tube of 1893 was a wide-bore semi-rigid orotracheal tube carrying an inflatable cuff and a pilot balloon to reflect pressure in the cuff.

The development of tubes and anaesthetic techniques has not been a linear progression, some concepts being 'rediscovered' more than once. In spite of Dorrance's description of the cuffed rubber orotracheal tube in 1910 – Guedel and Waters in 1928 described a similar tube for use with a carbon dioxide absorption technique – it was not until after the polio epidemic of the 1950s that the use of cuffed tracheal tubes became standard anaesthetic practice.

I.W. Magill and E.S. Rowbotham, anaesthetists to the British Army Plastic Unit in Sidcup during and after the Great War of 1914–1918, found that they could provide a superior unimpeded surgical field for the head and neck surgery of Sir Harold Gillies by having the patient breathing to and fro through an uncuffed rubber tube passed blindly through the nose into the trachea. Prior to this, 'insufflation anaesthesia' (as opposed to 'inhalation anaesthesia'), whereby chloroform or ether laden air was pumped into the trachea as advocated by Meltzer and Auer in New York,[59] had achieved rapid and widespread acceptance, partly owing to the different needs of thoracic surgery. This technique, when used for head and neck surgery, had necessitated a second tube being inserted for egress of gasses in order that the pharynx might be packed to prevent soiling of the trachea. It was this second wide-bore rubber tube (that Ivan Magill and Stanley Rowbotham quite by chance found they could 'blindly' insert into the larynx) which subsequently became the singular airway for 'to and fro' respiration and spawned the dominant technique in the UK for many years.

Of note, throughout this time the technique of laryngoscopy was still in its infancy. In 1941, Gillespie, writing in *Endotracheal Anaesthesia*,[54] stated: 'An experienced worker should be able to intubate all but the most difficult cases in ten minutes. The beginner will often require thirty.'

Design

An orotracheal tube (Fig. 6.23) usually has a preformed curve that approximately matches the anatomical curve of the airway. This aids insertion and ensures that when the tube is further flexed *in situ*, it is unlikely to kink. The distal end is cut obliquely (bevelled) so that when held in the right hand the aperture faces to the left. The bevel facilitates insertion and allows the tip of the tube to be seen passing between the vocal cords. There may be a hole (a Murphy eye) in the wall opposite the bevel. This is designed to provide a secondary port for gas movement in and out of the tube should the bevel become blocked or wedged against the tracheal wall (Fig. 6.24). Cuffed tubes,

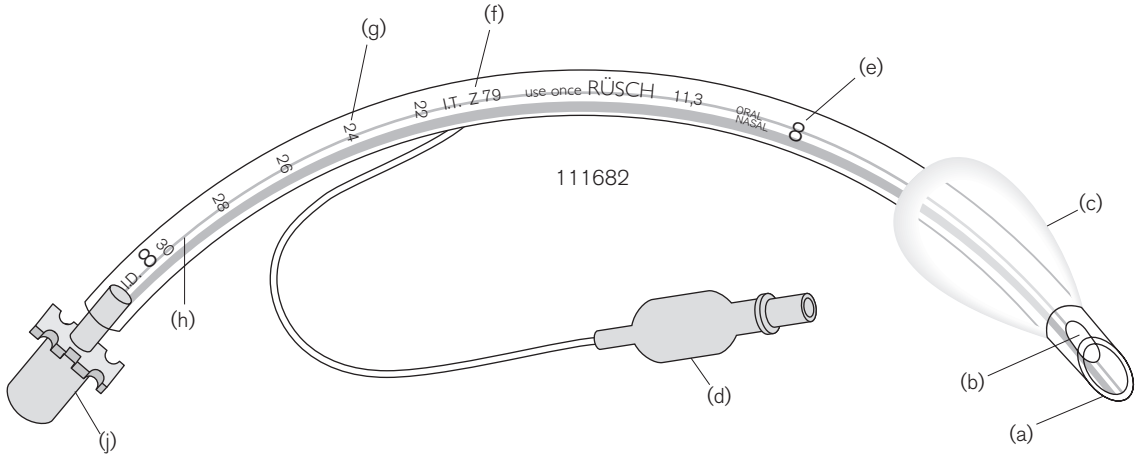


Figure 6.23 An uncut orotracheal tube. **A.** Bevel; **B.** Murphy eye; **C.** tracheal cuff; **D.** self-sealing valve which keeps gas in the cuff; **E.** marking to show the internal diameter of the tube in millimetres; **F.** marking (I.T.Z 79) to show that the plastic has been tested for tissue toxicity; **G.** the length of the tube in millimetres; **H.** longitudinal line of radio-opaque material; **J.** 15 mm connector.

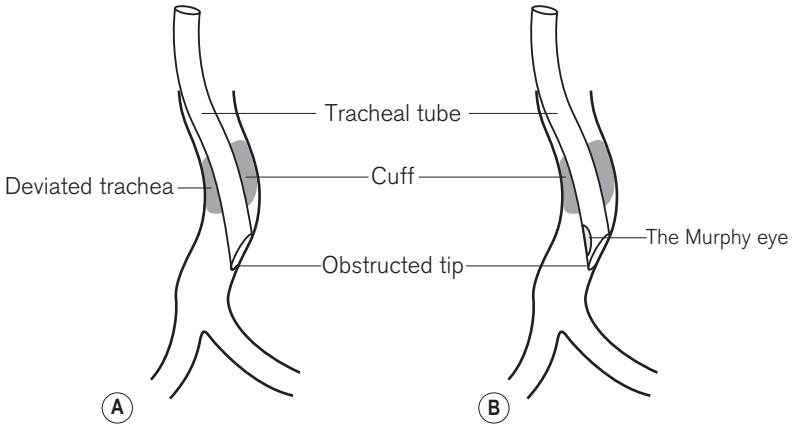


Figure 6.24 If there is marked deviation of the trachea, the distal orifice may become obstructed, a second orifice or 'Murphy eye' is then valuable.

by definition, have an inflatable cuff a short distance proximal to the tube tip, which, when inflated, seals the space between the tube and the tracheal wall.

Modern tracheal tubes carry several markings, one of which is a longitudinal line of radio-opaque material so that the correct placement can be verified from an X-ray if required. A transverse black mark on some tubes, made several centimetres proximal to the cuff, is designed to indicate the distance that the tube should be placed beyond the vocal cords; the mark should remain just visible above the larynx, indicating that the tube has not been inserted too far. The distance from the tip of the bevel is also marked in centimetres on the tube wall, along with the internal diameter (ID). Markings describing Implant Testing and CE markings are discussed below.

Construction materials

Although traditional red rubber tracheal tubes are now largely consigned to history, much of the design and conventions of modern tracheal tubes derives from their attributes, they therefore continue to be included here.

Tracheal tubes were previously made of red rubber or natural latex, which can be cleaned and sterilized for reuse. The material generally provides a combination of adequate rigidity and minimal mechanical tissue trauma, but it also has several disadvantages. As it is opaque, inadequate cleaning and foreign bodies within the lumen of the tube might be overlooked. Repeated sterilization causes degradation, with the potential for weakening leading to kinking or rupture. The materials themselves are potentially allergenic and irritant when used for long periods, being implicated in the development of laryngeal granulomata.

Currently, plastics (polyvinyl chloride (PVC) and more recently polyurethane) and to a lesser extent, silicone rubber, have replaced red rubber and natural latex as primary materials for the following reasons:

- They are non-irritant and are now inexpensive enough to allow single-patient use and hence sterilization at manufacture.
- As the material is usually clear, foreign bodies and blockages may be seen more easily.
- Manufacturing tolerances are much better with plastics, so that there is much less variation in the size of the lumen (most important in neonatal tracheal tubes).
- Cuffs of rubber tubes are usually thick and require high intracuff pressures to inflate them; inadvertent over-distension of the cuff may result in this pressure being transmitted to the tracheal mucosa with a risk of mucosal ischaemia or necrosis during prolonged use.

In contrast, plastic tubes do not have the elasticity (springiness) of rubber and may be less easy to insert in difficult situations. Their relative rigidity at room temperature, compared with rubber tubes, tends to increase trauma particularly when they are inserted via the nasal route. Strategies to minimize trauma include use of the smallest tube necessary, warming the tube in warm water before use or selection of a tube made of a softer compound (e.g. the Portex Ivory range).

The softness of plastic tubes is dependant on several factors. Polyurethane compounds tend to be softer and more springy than PVC, but are more expensive to manufacture. Chemicals called *plasticizers* may be added to PVC during manufacture to produce a softer product. The most widely used plasticizers are a group of chemicals called phthalates (particularly dioctyl phthalate in medical products). In recent years concern has been raised at governmental levels in both Europe and America about the effects of phthalates which may leech out of the plastic and some consider to be potentially "carcinogenic, mutagenic and reprotoxic".[2] While the level of risk posed by these additives is likely to be very low, several organizations seek their complete removal from manufacturing

Table 6.3 The classical view of the relative dimensions of tracheal tubes

INTERNAL DIAMETER (MM)		AGE (YEARS)	LENGTH (CM)	
Oral	**Nasal**		**Oral**	**Nasal**
2.5	2.5	PREMATURE	10.5	13.0
3.0	3.0	0–1	10.5	13.0
3.5	3.5		11.0	14.0
4.0	4.0	1–2	12.0	14.5
4.5	4.5		13.5	15.0
5.0	5.0	2–4	14.0	16.6
5.5	5.5		14.5	17.0
6.0	6.0	5–12	15.0	17.5
6.5	6.5	13–16	16.0	18.5
7.0	7.0		17.5	19.0
8.0	8.0		18.5	19.5
–	6.0	ADULTS	–	24.0
–	6.5	Small women Large men	–	24.0
7.0	7.0		–	24.0
7.5	7.5		–	25.0
8.0	8.0		23.0	26.0
8.5	–		24.0	–
9.0	–		25.0	–

A widely used formula for selecting the diameter of an tracheal tube suitable for children over the age of 1 year is:

$$\frac{\text{Age in years}}{4} + 4.5\,\text{mm}$$

The exact length to which a new tube should be shortened cannot be categorically specified. In some operations it is necessary to pass the tube further down the trachea than in others. Cuffed tubes are generally trimmed to a centimetre or so longer than plain ones.

the trachea considerably (1 mm of circumferential swelling reduces the diameter of the adult airway by less than 10% and that of an infant by as much as 30%) and may lead to respiratory compromise. Because of these considerations paediatric tracheal tubes have traditionally been uncuffed and use of a tube small enough to leave an audible leak is still standard. Use of uncuffed tubes leads to an unpredictable leak so children have routinely been ventilated with pressure-control ventilation (ensuring ventilation despite a variable leak), while volume-controlled ventilation has been favoured in adults (delivering an assured minute volume despite changes in lung compliance).

While use of uncuffed trachea tubes in paediatrics remains the norm, technology and changing circumstance has led to an increase in the use of cuffed tubes (see below).

Tracheal tube cuffs

High-pressure cuffs

In the era of red rubber tracheal tubes the requirement to sterilize the tubes by autoclaving for reuse necessitated construction that was strong and resilient. As a result tracheal tube cuffs were made of a relatively low compliance thick rubber. Repeated cleaning and reuse could lead to either hardening of the cuff making it ever more rigid or, occasionally, the development of weaknesses which if overinflated could 'herniate' over the tip of the tube and occlude it (Fig. 6.25). These cuffs required a high pressure to distend them and were relatively low volume (high-pressure low-volume cuffs). Thick-walled cuffs also tend to inflate in a circular shape rather than conforming to the shape of the (non-circular) trachea within which they lie.

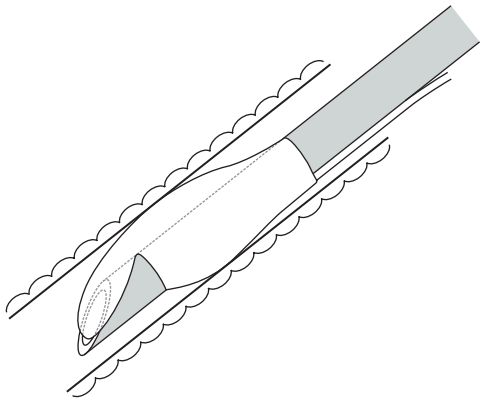


Figure 6.25 Herniated cuff.

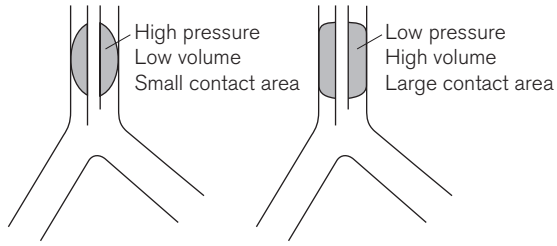


Figure 6.26 High- and low-pressure tracheal cuffs. Medium-pressure cuffs have a profile between these extremes.



Figure 6.27 Large-volume cuff inflated in a model trachea shows formation of numerous folds and channels. After Latto 1997.[66]

In order to achieve enough contact with the tracheal wall and a good seal relative over-inflation was required, with the result that the high pressure within the cuff was transmitted to the tracheal wall. This readily led to a reduction of mucosal pressure to critical levels (capillary perfusion pressure is usually about 35 mmHg) and if prolonged could lead to mucosal ischaemia which in turn in a proportion of patients caused the development of tracheal scarring and tracheal stenosis.

Medium-pressure cuffs

These are made from a much thinner elastic material such as latex rubber which fits snugly to the tube in its deflated state without appearing too bulky. The Intubating LMA tracheal tube is an example. An intermediate amount of pressure is required for inflation, but because of the lower intra-cuff pressure and the material's greater deformability it adapts better to the shape of the tube it lies within (the trachea) and seals without interfering with tracheal mucosal perfusion. Due to the greater compliance of the cuff material, over-inflation is also less likely to lead to dramatic pressure rises and subsequent harm.

Low-pressure or high-volume cuffs

These are made from a thin inelastic material (PVC) which when fully inflated would have a larger volume than that required to effect a seal (Fig 6.26). *In situ* there is a large area of contact between the cuff and tracheal wall before full inflation of the cuff. The pressure within the cuff can therefore be kept much lower and can achieve a seal with minimal risk of occluding mucosal blood flow. Both medium- and low-pressure cuffs are made of very thin materials which may be ruptured by over-distension or more often damaged by snagging on teeth, instruments or passage through the nose. Low-pressure designs now predominate. This is particularly so in tubes such as tracheostomy tubes used for long-term ventilation on intensive care units.

One drawback of the low pressure cuff is that because the cuff material is not fully unfolded when a seal is achieved, a number of small folds (micro-folds) can create small channels (micro-channels) running the length of the cuff. These channels may contribute to the causation of ventilator-associated pneumonia (VAP) by allowing transmission of potentially infective pharyngeal contents past the cuff. It is readily demonstrated *in vitro* (Fig 6.27) that these micro-folds are not waterproof and despite the mantra of the tracheal tube 'protecting' the airway it must be acknowledged that this protection is incomplete. There

is increasing interest in the role of micro-aspiration in VAP amongst patients on intensive care.

Polyurethane cuffs may be made that are even thinner than PVC cuffs and recently this has contributed to renewed interest in the use of cuffed tracheal tubes for children. The technology now exists to make cuffs thin enough to not interfere with the insertion of neonatal size tracheal tubes, yet strong enough for routine use. While such use is relatively uncommon it has its proponents, citing both increased reliability of ventilation, and increased protection of the lower respiratory tract, together with the facility for using smaller tubes which exert their seal pressure below the narrowest part of the paediatric airway.

Tracheal tube (and SAD) cuffs are inflated via a small-bore inflation tube (pilot tube) either welded onto the outside of the tracheal tube or more commonly now integrated into its wall. The inflation tube is connected, at its proximal end to a small 'pilot balloon' to give an indication of the distension of the cuff. Accessing the pilot balloon by a one-way valve allows inflation of the device's cuff.

All cuff types should only be inflated to the pressure necessary to achieve sufficient seal to prevent gas leak at the airway pressures generated by positive pressure ventilation. This has the added advantage that it permits leakage (often audibly) if there is overpressure in the breathing system. During prolonged use of a tracheal tube intermittent use of a manometer attached to the pilot balloon allows maintenance of an intra-cuff pressure below 30 cm H_2O. Only one manufacturer currently makes integrated pilot pressure monitors (Cuff Pilot; see above, 'Other Laryngeal Masks') for tracheal tubes, though further development is likely.

Nitrous oxide and tracheal tube cuffs

Nitrous oxide (N_2O) diffuses into tracheal tube cuffs filled with air. This applies also to cuffs on supraglottic airways. The rate at which diffusion occurs depends on:

- the permeability of the material from which the cuff is constructed (rubber being more permeable than plastic)
- the surface area of the cuff exposed to N_2O
- the partial pressure of N_2O in the respiratory gas.

The rise in intra-cuff pressure caused by this diffusion into the cuff depends on the compliance of the cuff. Low-volume high-pressure cuffs suffer the greatest pressure rise and may well transmit this pressure rise to the tracheal mucosa. High-volume low-pressure cuffs expand with only a slight increase in pressure until full inflation; at this point, owing to the inelasticity of the material, the pressure may rise rapidly to up to 12 kPa (90 mmHg). This can damage the tracheal mucosa.[67] In practice equilibration takes place over approximately 20 minutes, so checking the cuff pressure after 20 minutes will minimize problems. Should the anaesthetic gas mixture be changed at a later time to increase (or decrease) the fractional concentration of nitrous oxide, further diffusion into (or out of) the cuff will occur.

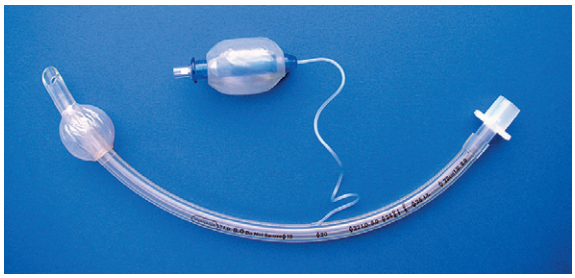


Figure 6.28 The Mallinckrodt Brandt device on a size 8 tracheal tube.

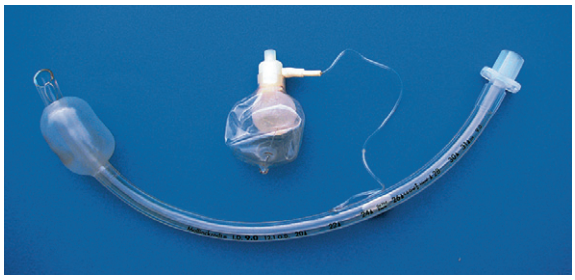


Figure 6.29 The Mallinckrodt Lanz device on a size 9 tracheal tube.

This phenomenon can be avoided if the cuff is filled with either a liquid, or a gas mixture identical to that of the inspired gas, or else by regular monitoring of the cuff pressure and volume. Alternatively, there are several devices that limit the pressure rise:

- The Mallinckrodt Brandt device has a large pilot balloon made from a material that allows nitrous oxide which has diffused into the cuff to escape (Fig. 6.28).
- The Mallinckrodt Lanz (Fig. 6.29) device consists of a special control valve and pilot balloon arrangement. A compliant latex balloon, protected within a larger open plastic covering, connects to the cuff via a flow restrictor. Insertion of a syringe into the valve bypasses this restriction allowing rapid inflation of the cuff. Thereafter, gradual volume changes in the cuff are buffered by the pilot balloon. When filled with a preset volume the device regulates the cuff pressure to between 20 and 25 mmHg.
- Portex Soft Seal cuffs are made of a PVC material using a higher molecular weight plasticizer, which results in a five-fold lower permeability to nitrous oxide. Fig. 6.30 compares rises in intracuff pressure between a Soft Seal and two other cuffs

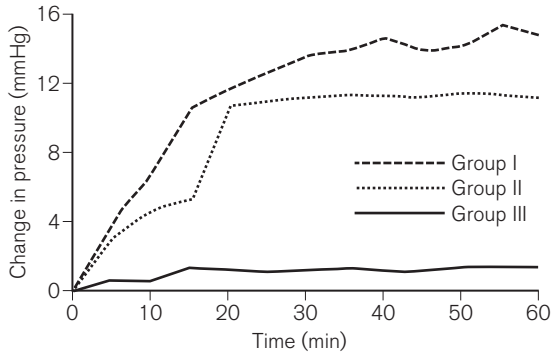


Figure 6.30 Mean changes in intracuff pressure every 5 min in three groups: group I, Mallinckrodt Lo-Contour tube; group II, Portex Profile tube; group III, Portex Soft Seal tube. *From Al-Shaikh* et al *1999,*[68] *with permission of Oxford University Press.*

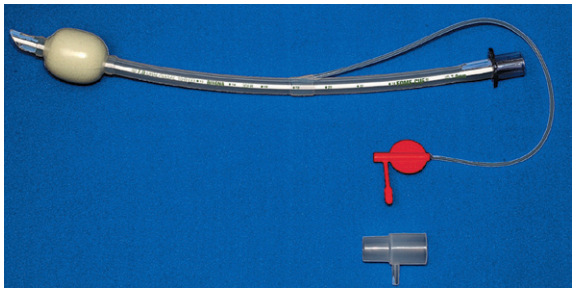


Figure 6.31 Bivona 'Fome Cuf' tracheal tube. The pilot tube for deflating the cuff can be connected into the ventilation circuit to equilibrate intracuff pressure during IPPV.

during 1 hour of anaesthesia using 66% nitrous oxide.

- The Bivona 'Fome-Cuf' uses a piece of shaped foam to replace the air within the cuff. Prior to insertion the cuff is evacuated to collapse the foam and after insertion the pilot tube is opened to the atmosphere to allow the foam to expand, to 'inflate' the cuff (Fig. 6.31). Because the pilot tube is left open, diffusion of nitrous oxide does not further inflate the cuff. The pilot tube may also be connected to the ventilation circuit to equilibrate intracuff pressure with proximal airway pressure during positive-pressure ventilation.

Nasotracheal intubation

The diameter of tubes inserted through the nose is clearly limited by the size of the nares and the air passages of the nasal cavity. The position of the nasal septum and turbinates may result in a fairly convoluted passageway for intubation. Smaller diameter, softer and thinner-walled tubes are therefore needed for ease of insertion and to diminish the risk of bleeding. Similarly, a streamlined cuff made of material that is not easily torn is preferable. The pilot tube is incorporated as far as possible in the body of the tube.

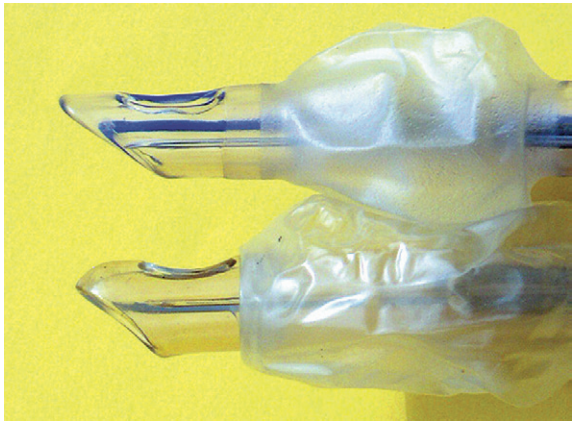


Figure 6.32 A blunt cupped tracheal tube bevel, compared with a standard bevel.

The flexible LMA has removed the need for nasal intubation for many cases in which it has been traditionally used. In some centres uncuffed nasal tubes remain popular for use in adults using spontaneously breathing techniques, such as for short dental cases. The larynx is packed to protect the lungs for the period of the surgical procedure, and afterwards the small soft tubes cause minimal irritation and can be removed once the patient has woken. The necessity to use the highly error prone throat pack is a potential disadvantage. They increase the incidence of sore throat but, more importantly, in spite of many years of use anaesthetists and surgeons still contrive to devise seemingly new and exotic circumstances and procedures for forgetting to remove them in order to cause postoperative airway obstruction.[69]

Specially designed nasal tubes which are made of soft plastic, are preformed and incorporate an appropriate cuff are probably best suited for such techniques. Designs such as the north-facing Portex polar tube (see below) enable safe fixation and removal of the anaesthetic circuit from the face.

The design of the bevel is also important; sharp long bevels have tended to be replaced by short bevels with rounded tips (Fig. 6.32) which are less likely to snag on obstructions. Some manufacturers even use a different softer compound for the leading edge of the tracheal tube.

Common problems with the use of tracheal tubes

Many of the problems, particularly relating to tube kinking or obstruction by external compression, are now rarely

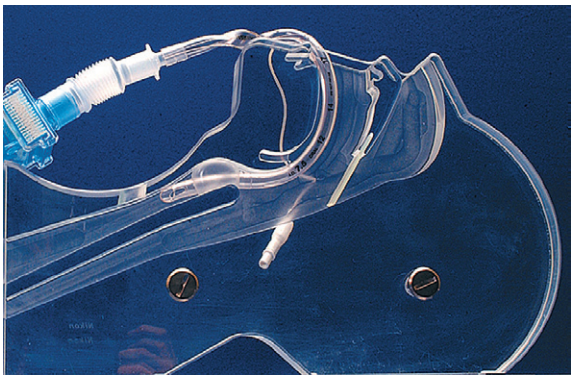


Figure 6.33 Occlusion of the tracheal tube due to rotational forces by the breathing system.

seen with the use of PVC tubes. These soften with warming when in use, appearing more often to remould rather than kink abruptly: so much so that specially designed non-kinking tubes, such as the Oxford tube and even the 'armoured tube' are now largely obsolete (see below). Some of the common problems are:

- *Disconnection* of the tube from the 15 mm ISO connector or of the connector from the catheter mount is common. This is most likely during shared airway or head and neck surgery. With the patient's head often positioned away from the anaesthetist and hidden by drapes and surgeons, vigilance and appropriate monitoring (which includes alarm-enabled capnography) is vital.
- *Leak.* In uncuffed tubes excessive leak indicates a larger tube is required. With cuffed tubes a leak may arise if the cuff is at the cords. It may also indicate inadequate cuff inflation or deflation caused by cuff damage, over-inflation or a fault in manufacture.
- *Kinking external to the patient*. Tubes are prone to being kinked by the position of the breathing system, e.g. an uncut tube emerging from the mouth or as in Fig. 6.33, due to rotational forces being applied by a breathing system. Paediatric tubes, which have thinner walls and are connected to relatively heavy attachments, are particularly prone to this, especially when softened by warm respiratory gasses.
- *Kinking inside the patient* (Fig. 6.34). Modern PVC tubes are fairly resistant to this form of obstruction. If a tracheal tube must be bent acutely when *in situ*, a reinforced tube or one that is specially shaped should be used (see below, Tubes for special purposes). When the airway is shared with the surgeon, tube kinking and extubation or disconnection are increased problems. The Boyle-Davis gag used for tonsillectomy, if used without care or experience or with the incorrect blade size, frequently causes kinking (and/or inadvertent extubation on removal, and can advance an overly long tube into the bronchus). Obstruction by biting is a risk as a patient emerges from anaesthesia. Red rubber tubes were prone, especially after repeated autoclaving, to obstruction by a tightly inserted throat pack or even by inward herniation of the tube wall underneath the high-pressure cuff.
- *Foreign bodies.* All manner of foreign bodies have been found within airway devices causing blockage. Tracheal tubes (and other airways) should not be placed in the same dish as Luer-Lock caps, needles and ampoules etc., even after use. Diaphragms of dried mucus and even K-Y jelly, which may be virtually invisible, have also been found blocking tubes. In single-use devices, this danger has been superseded by diaphragms of clear plastic being formed across the openings of tubes when, for unwrapping, they are pushed through their plastic packaging. The invisible film is then kept firmly in place by the connector when the next item in the sequence is attached. Rarely manufacturing defects

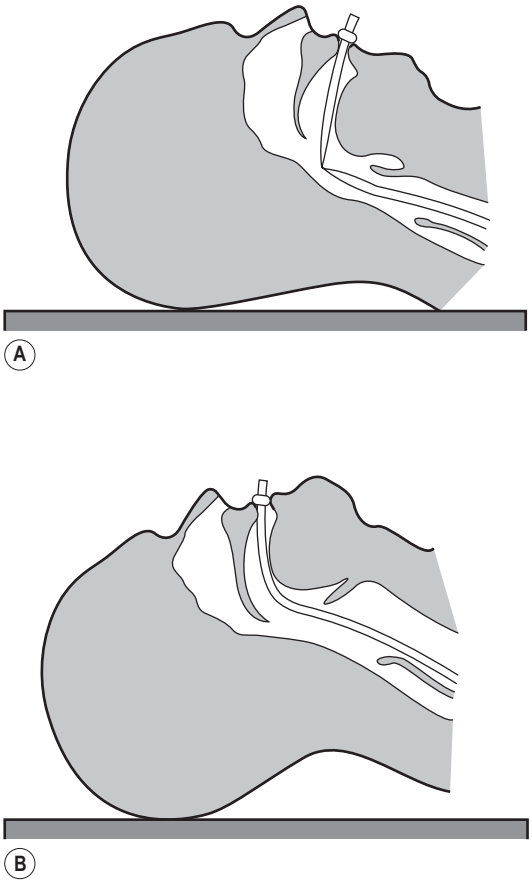


Figure 6.34 **A.** Kinking of an tracheal tube inside the mouth following neck flexion and softening of the tube during anaesthesia. **B.** Head in correct position.

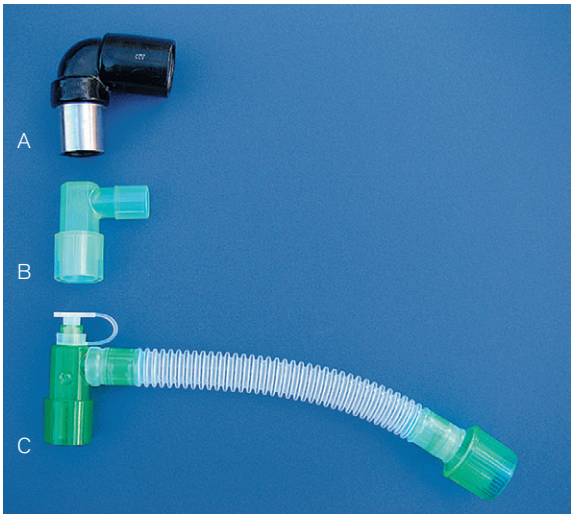


Figure 6.36 Angle pieces: **A.** older version with 22 mm tapers; **B.** newer design (here single-use) with proximal 15 mm external taper and distal dual mode taper (external 22 mm male and internal 15 mm female); **C.** combined angle pieces and catheter mount arrangement (see text).



Figure 6.37 A variety of tracheal tube connectors. Clockwise from top left: intersurgical flexible corrugated models for gas sampling, and with 'Seal-Around' cap on swivel mount for bronchoscopy or suction; Mainz Universal Adaptor (blue-coloured) for endoscopy/intubation via tracheal tube or facemask; plain angle piece; swivel mount.

could be attached to an airway device, such as the Boyle-Davis gag or to a catheter that was passed through the larynx into the trachea. In the latter case, the gasses and/or vapours were blown constantly down the catheter and were exhausted to the atmosphere by passing back through the trachea but outside the catheter (see above, under History).

Currently, a catheter mount usually consists of a short piece of flexible kink-resistant 15 mm tubing with a 15 mm ISO female tapered connector for attachment to the tracheal tube (Fig. 6.38). The other end of the catheter mount may have either a 15 mm ISO male taper or a 22 mm ISO female taper connector for coupling with the breathing system. Although the device increases the apparatus dead space of the breathing system, the flexibility it imparts makes this an insignificant consideration except in the very young.

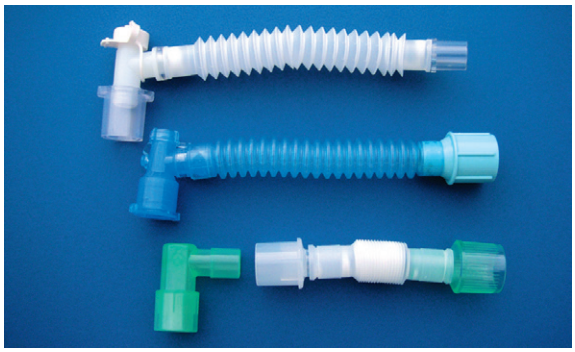


Figure 6.38 A selection of catheter mounts, showing the 15 and 22 mm ISO connections described in the text.

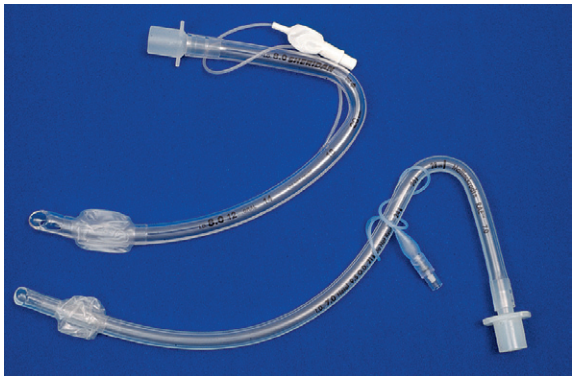


Figure 6.39 RAE design tracheal tubes for orotracheal intubation (top) and nasotracheal intubation.

Tracheal tubes for special purposes

Many 'special' tracheal tubes have been devised of which only a few examples are described below. Manufacturers' designs of the same type of tube may vary considerably with even small variations having significant effects on the practicality of devices; the 'same tube' provided by different manufacturers may vary in a number of features, be it tip design (e.g. Murphy or Magill type), cuff and pilot balloon design, or even construction material or additional lumens. Some designs, however, are 'branded' and available only from one company, although others may have a similar design under a generic description. For a full range of tubes or for particular requirements, it is necessary to consult the manufacturers directly.

RAE preformed tubes

There are two separate versions of these tubes, named after their inventors: Ring, Adair and Elwin (Fig. 6.39). The tube

for oral intubation has a preformed bend on the part of the tube where it exits the mouth, so that proximal tube passes down the chin and away from the face, thus improving surgical access to the mouth, nose and head of the patient. The nasal version is bent where it exits the nose so that the part housing the ISO connector passes upwards to the forehead. It is used to improve surgical access for intra-oral procedures.

The main disadvantage of both designs is that the bend in the tube determines the length of tube that lies within the patient. As patients inevitably vary in size the tube may be too long for a given patient leading to the tip entering the bronchus; this is a particular problem with small children. Conversely, and much more commonly, with the trend towards using smaller calibre tubes, because tube dimensions have not been reformulated, a chosen RAE tube may be too short to reach comfortably below the glottis, increasing the risk of inadvertent extubation (see above: Tracheal tubes, Size). This is a particular problem with the newer cuffed paediatric tubes where manufacturers appear to have simply added a cuff to existing tubes rather than reformulating the proportions.

Reinforced tubes

Most tracheal tubes will kink if bent into an acute enough angle or if compressed by an external force such as from a surgical instrument. Both can occur during the course of an operation. Tubes may be made kink-resistant (but not kink-proof!) by embedding a reinforcing spiral of steel or nylon wire into the wall of the tube, which can then be made of a more elastic material than usual (i.e. silicone rubber, latex rubber or soft plastic). The tubes are consequently very flexible with little preformed shape. This can make insertion into the trachea more difficult, often requiring either a bougie, which has first been passed into the trachea, or a malleable stylet held in the lumen to shape the tube. Reinforced tubes have a thicker wall and therefore a smaller internal diameter (ID) for a given outer diameter.

These tubes are usually supplied at a length of 30 cm or greater for an ID of 5.0 mm and above. The reinforcing spiral will not stretch to accommodate a connector so the tubes cannot be cut to a shorter length than supplied. The ISO connectors are therefore usually bonded at production to a non-reinforced part of the tube. Special care must be taken not to insert these tubes too far. Reusable latex versions were particularly prone to kinking at the 'soft spot' between the end of the connector and the start of the spiral reinforcement (Fig. 6.40).

The greater kink resistance of PVC tubes and the availability of polar tubes has diminished the need for using reinforced tubes. Furthermore, the metal wire spiral may obscure radiolological imaging in some cervical spine surgery.

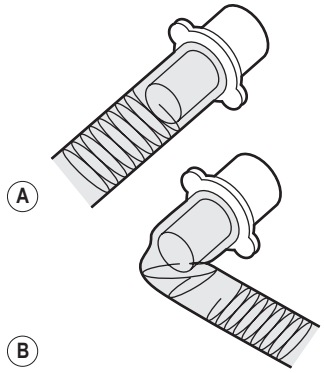


Figure 6.40 Obstruction at the 'soft spot' between the connector and the spiral of an armoured tube. **A.** With the connector placed correctly; **B.** with incorrect placement leaving an unarmoured segment of tube.

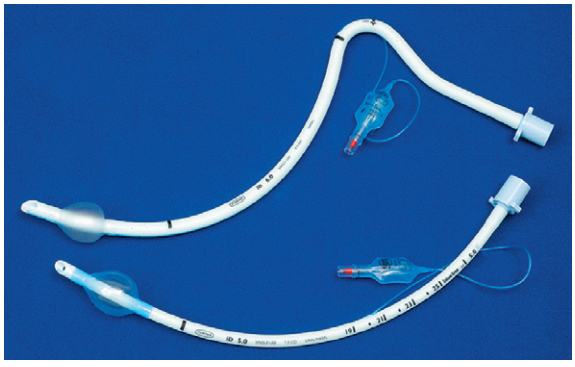


Figure 6.41 Portex Ivory tracheal tubes. Upper: north facing directional (polar) tube. Lower: microlaryngeal tube with greater length and cuff size relative to the tube diameter.

Polar tubes

Portex (Smiths Medical), in their Ivory range of tubes, manufacture a tube for nasotracheal intubation in a soft PVC material (see Tracheal tubes, Materials) with a preformed curve that takes the proximal end of the tube snugly along the nose and over the forehead (Fig. 6.41). These tubes are well-suited to maxillofacial surgery as they impinge relatively little on the surgical field. If access is also needed to the forehead, as in some cosmetic or reconstructive plastic procedures, the proximal end can be swung down without the tube kinking, to descend over the cheek or chin. The soft plastic of the Ivory range is relatively atraumatic in the nose, which in combination with the springiness of these tubes makes them ideal for 'blind nasal intubation'. Interestingly, the material, originally called 'Vinyl Portex Tubing', was actually first developed in 1944[66] as a substitute for the red rubber tubes whose primary materials were in short supply.

These tubes, in sizes 6.0 and 6.5 mm ID, are a good choice for nasal fibreoptic intubation, in spite of the reservations previously mentioned regarding cuff sizes and small tubes (the length and cuff size in this case is usually adequate, even for the largest adults).

Microlaryngeal tube

This tube (Fig. 6.41, lower) has a small internal diameter (ID usually around 4.5–5.5 mm) but an adult-sized high-volume low-pressure cuff. The tubes are designed to be unobtrusive in the larynx so as to allow surgery around the vocal cords and are long enough for nasal intubation. The high resistance to gas flow in these tubes virtually obligates controlled ventilation and a long expiratory phase should be used to allow complete expiration. Due to their flexibility they can be difficult to insert unless Magill intubating forceps are used. Standard adult-sized bougies and stylets are usually too large to pass through these tubes. Despite their suitability for laryngeal surgery they tend to lodge in the anterior of the laryngeal inlet so do not provide surgical access for all laryngeal lesions. As they are not laser resistant they are unsuitable for such surgery.

Carden tube

This tube was developed to facilitate microsurgery of the larynx. It is rarely used now but may perhaps enjoy a resurgence for jet ventilation. It comprises a shortened cuffed tracheal tube that sits wholly below the glottis attached to a long catheter for insufflation of gas and a long pilot tube for the cuff (Fig. 6.42). It may be inserted by grasping the tube with Magill's forceps and placing it under direct vision. Alternatively the Carden tube and an uncut plain tube, just wide enough to fit inside it, are threaded onto a stylet. This assembly is introduced through the larynx so that the Carden tube and a centimetre or so of the plain tube pass into the trachea. The cuff of the Carden tube is inflated, the stylet is withdrawn and anaesthesia is maintained in the usual manner through the plain tube. For laryngoscopy, the plain tube is removed as is the breathing system, and gasses are delivered directly to the Carden tube through a feed mount. The expired gasses escape via the lumen of the Carden tube.



Figure 6.42 Carden tube for laryngeal surgery.
Photo courtesy of Portex UK Ltd.

Tubes for laser surgery

Conventional tracheal tubes of either plastic, silicone or rubber may be damaged by the carbon dioxide, KTP or Nd-YAG laser beam. These materials burn more fiercely in the presence of oxygen or nitrous oxide than in air and are easily ignited by a direct or indirect strike from the laser beam. The resultant fire can produce serious upper airway burns and severe distal inhalation injury; injuries may be fatal. Some examples of tubes that can be used in the presence of a laser beam are described below. They must not be assumed to be laser *proof* and are only indicated for CO_2 and KTP lasers. Figures are provided by manufacturers for maximum power and duration of energy before ignition for different lasers according to standardized tests (e.g. 1 s at 55 watts or 7 s at 30 watts from a CO_2 laser for the Laser-Flex).

Mallinckrodt 'Laser-Flex' tracheal tubes

These single-use tubes for oral intubation are made from a gas-tight metal helix with the pilot tubes for the double cuff carried within the lumen of the airway (Fig. 6.43). It is advised that the cuffs, which are not laser resistant, are inflated with saline, ideally dyed with some methylene blue, to prevent ignition of the cuff and so that puncture is obvious. The function of the proximal cuff is to protect the distal cuff and the tube must be replaced if either cuff is defective. The surface of the tube shaft reflects a defocused laser beam with a lower potential for causing tissue destruction. The tip of the tube is made of a soft plastic for a more atraumatic insertion. The tubes have a markedly reduced inner diameter when compared to PVC tubes



Figure 6.43 Mallinckrodt 'Laser-Flex' tracheal tubes. Uncuffed and cuffed showing water-filled double cuff arrangement.

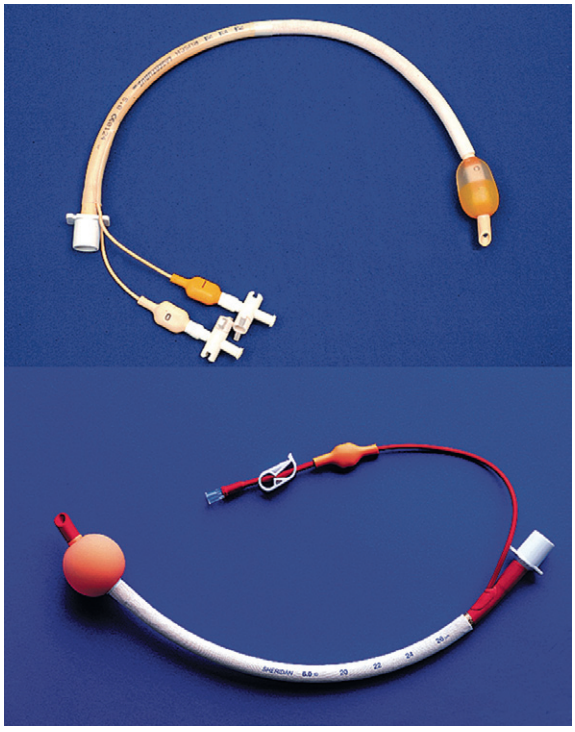


Figure 6.44 The Rusch Lasertubus (above) and Sheridan Laser-Trach (below, photo courtesy of Hudson RCI) laser resistant single-use tracheal tubes. The embossed copper foil of the Sheridan tube is just visible under the outer covering.

of the same outer diameter and care must be taken to allow adequate expiratory time in ventilated patients.

Foil wrapped

An alternative approach used by some manufacturers is to spirally wrap a suitably small rubber tube with a narrow strip of silver or copper foil (this may be stippled or corrugated to disperse an incident beam) (Fig. 6.44). The tube is then overwrapped to give a smooth outer coating. The Rusch Lasertubus is covered in Merocel foam (a porous sponge), which is soaked in saline to help absorb laser energy. This tube is provided with a double cuff arrangement.

Jet ventilation

This provides a further approach to airway management for laser surgery and is described separately below.

Tubes and catheters for 'jet ventilation' (high-pressure source ventilation)

Supraglottic ventilation

If surgical access is significantly impaired by a laser resistant tube, the surgical laryngoscope that is used to suspend the larynx for surgery can be fitted with a metal cannula

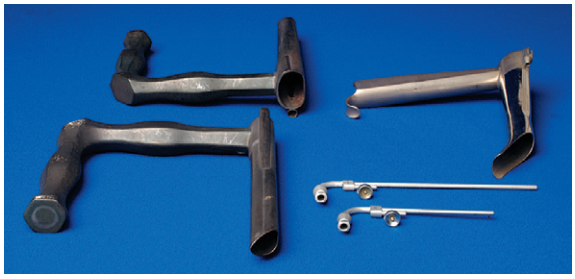


Figure 6.45 Different patterns of laryngoscopes for suspension surgery and examination of the larynx, and alongside, jet ventilation cannulae for use with these devices.

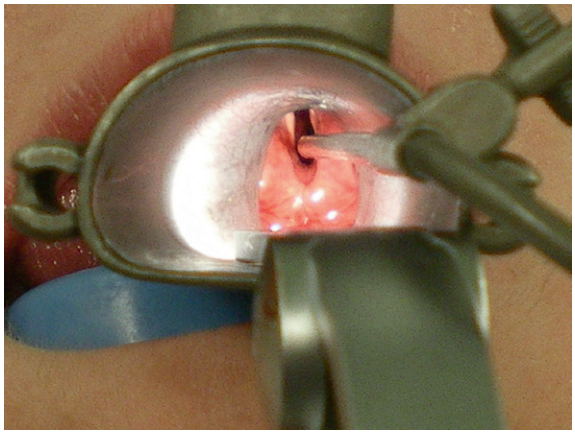


Figure 6.46 Supraglottic jet ventilation with cannula correctly aligned for effective ventilation.

(Fig. 6.45) that ends above the glottis and can be used for 'jetting' with high-pressure gas (*supraglottic ventilation*). Anaesthesia must be maintained intravenously. The quality of ventilation depends on the surgeon placing the tip of the ventilating catheter above and in line with the trachea and requires careful co-operation between anaesthetist and surgeon (Fig. 6.46). Alternatively, high-pressure source ventilation may be delivered by a small laser-resistant catheter passed through the glottis (*transglottic ventilation*).

Transglottic ventilation – Hunsaker tube

The Hunsaker Mon-Jet Ventilation tube (Fig. 6.47) (Medtronic Xomed, USA) is a narrow-bore dual lumen tube designed for transglottic high-pressure source ventilation that allows concomitant airway pressure monitoring or gas sampling from the narrower second lumen. The 33-cm-long device (4.5 mm at its widest diameter) is made of a laser-resistant fluoroplastic and ends in a moulded plastic cage which aims to centralize the discharged jet stream and prevent direct impingement on the tracheal wall mucosa. In combination with a dedicated jet ventilator, such as the Accutronic Medical Systems

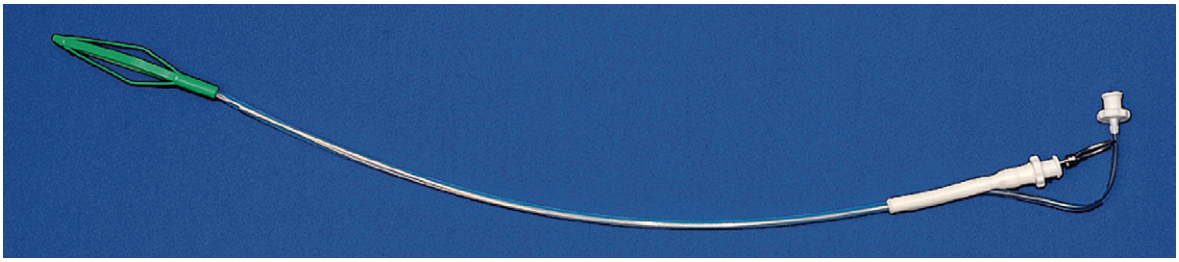


Figure 6.47 The Hunsaker Mon-Jet Ventilation tube.

Monsoon or Mistral models (see Chapter 9), this device offers improved surgical access for unhurried instrumentation and laser surgery of the major airways. Unlike the microlaryngeal tube the Hunsaker tube can be easily positioned to allow the surgeon access to any part of the larynx. Because the above ventilators check that airway (pause) pressure is below a pre-set limit before each inspiratory phase even at high cycle rates, jet ventilation can comfortably and safely continue up to and beyond patient arousal at the end of surgery.

Although the material will not melt and drip or produce a self-sustaining flame, protracted impact of sufficient laser energy will penetrate the tube. An internal stainless steel wire is designed to retain the distal fragment in the event of the tube being severed. Other manufacturers produce similar designs.

Subglottic ventilation: transtracheal catheters A cannula placed percutaneously into the trachea via the cricothyroid membrane enables *subglottic ventilation* while allowing the surgeon full access to the larynx. The Ravussin catheter ideally designed for this purpose is described below under Subglottic Devices.

The use of small tubes requires a high pressure gas source for ventilation. This may be a manually operated, fixed-pressure ventilator (Sanders injector, 4 bar); a manually operated, variable-pressure ventilator (Manujet, 0.5–4 bar; see below) or an automatic ventilator. Mechanical ventilators may be of a variety of degrees of sophistication and may provide conventional ventilation, high-frequency ventilation or a combination of the two. The use of a high-pressure source introduces a significant risk of barotrauma (particularly with manual techniques) and this risk increases if the larynx closes or if expiration (which occurs passively through the native airway) is not scrupulously confirmed before each inspiration.[70,71] Ventilators with airway pause pressure monitoring linked to ventilator cut-outs (see above) are safer and their use is recommended.[72]

Laryngectomy tube

Conventional tracheal tubes placed through a tracheostomy in patients undergoing or with previous laryngectomy are difficult to fix in place and prone to kinking or slipping down into a bronchus. The 'U' shape

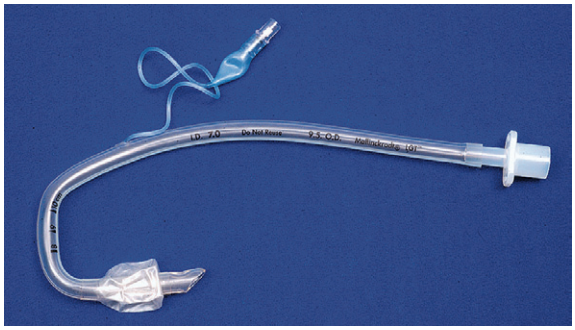


Figure 6.48 A laryngectomy tube.

of the laryngectomy tube (Fig. 6.48) means it sits unobtrusively on the chest wall with the long straight proximal portion allowing direct connection to a breathing system distant from the operative site. The internal portion of the tube is short and the tip is usually cut close to the cuff to minimize the risk of bronchial intubation.

Tubes for thoracic surgery (including bronchial blockers)

Operations within the thoracic cavity may require the collapse of a lung to improve access to other organs, to isolate that lung for its removal or repair, or to prevent contamination from a diseased lung spreading to the other side. This may be achieved in one of three ways.

(Endo)bronchial tubes

Firstly a long tube may be passed, often under endoscopic control, beyond the carina and into the unaffected lung so that the cuff can be inflated within a main bronchus. The other lung is not ventilated and collapses, especially when the pleura is opened. This technique is normally applicable only to intubating the left main bronchus (for surgery on the right lung). This is because the right upper lobe bronchus enters its main bronchus very close to the carina and its lumen would be obstructed by the cuff of the tube described. The bronchial tube may be placed in the trachea initially so as to ventilate both lungs for as long as possible and then advanced when required to isolate the appropriate lung.

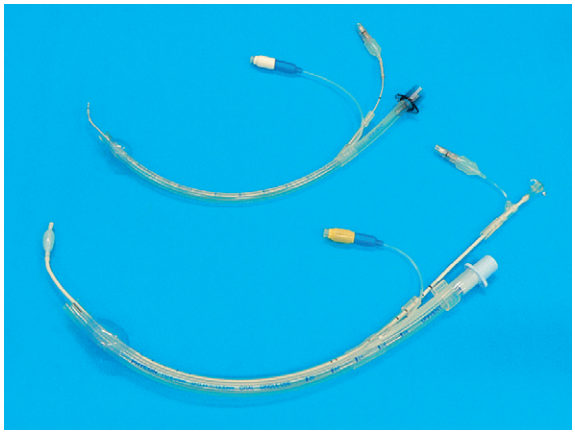


Figure 6.49 Univent tracheal tubes with endobronchial blockers. The lower device allows access to the lung distal to the blocker.
Photo courtesy of Fuji Systems Corporation.

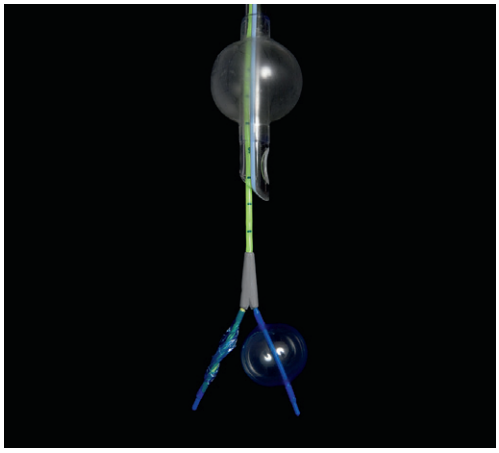


Figure 6.50 EZ-Blocker (AnaesthetIQ BV, The Netherlands) for lung isolation, deployed through a conventional tracheal tube.
Image courtesy of Fannin UK.

Tubes with bronchial blockers

The second method of isolating a lung is to use a tracheal tube through which a fine balloon catheter can be passed either via the lumen of the tube or through a channel in its wall. The balloon tip is then passed, usually with the aid of a flexible endoscope, into the intended main bronchus, which becomes isolated when the balloon is inflated (Fig. 6.49). A central channel which opens beyond the balloon allows suction to deflate the lung and may also be used to augment oxygenation by 'insufflation' into the collapsed lung. In the past Foley catheters and Fogarty embolectomy catheters have been used in place of bronchial blockers but improved construction of bronchial catheters obviates this need.

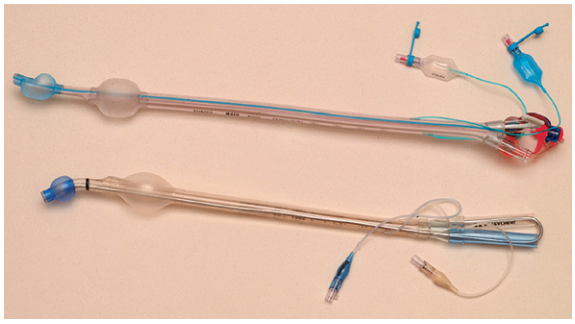


Figure 6.51 Double lumen tubes (both with malleable tracheal introducers in place). Portex right-sided tube (above) and Mallinckrodt Broncocath left-sided tube (below). The scalloping of the distal cuff is just visible on the right-sided tube (see text).

Bronchial blockers have a variety of designs to assist placement; these include fixed angulations, loops for attachment to an endoscope and a novel design with dual balloons either side of a self-deploying 'V' (Fig. 6.50). In this latter design the blocker is introduced within a sheath to the carina then advanced beyond the sheath, so the 'V' opens up and inevitably straddles the carina, with the dual balloons enabling deflation of either lung as required.

Blocker technology has improved considerably in recent years and, as evidence suggests their use is less injurious to the patient than use of a double lumen tube, popularity of blockers has increased, while that of double lumen tubes has waned. Blockers are now available for use even through infant-size tracheal tubes.

Double lumen tubes

The third method is to use a combination or double lumen tube (DLT) (Fig. 6.51). These are formed by bonding together two tubes of similar diameters but different lengths. The shorter tube ends in the trachea and seals this with one cuff, whilst the longer tube is designed to fit a main bronchus and seal this with a separate cuff. The cuffs are supplied by separate inflation tubes and pilot balloons, which are marked and colour-coded to aid identification. DLTs are produced in left- and right-sided versions. Some designs have a small soft carinal hook to prevent over-insertion of the tube but this is uncommon in modern single-use versions. Because the opening of the right upper lobe bronchus is so close to the origin of the main bronchus, the bronchial lumen on right-sided versions also has a side hole and a cuff that is designed so that it does not occlude this (Fig. 6.52). The bronchial cuff and inflation tube are blue by convention.

Sizes

The combined size of the two lumens makes the device bulky and prevents its use in children. Because the external shape of the device is circular in cross-section, the lumens

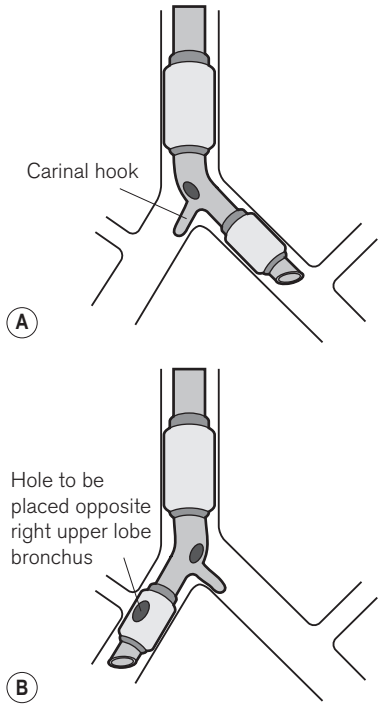


Figure 6.52 **A.** The endobronchial section of the tube in the left main bronchus with the carinal hook limiting the depth of insertion. **B.** The endobronchial section of the tube in the right main bronchus with the carinal hook limiting the depth of insertion and the side hole opposite the right upper lobe bronchus surrounded by the adapted bronchial cuff.

of the two tubes are each 'D'-shaped and the size cannot be quoted in terms of an internal diameter. Sizes are quoted in French gauge (Fr). This scale is constructed from the external diameter of the widest part of the device, measured in millimetres, multiplied by a factor of 3 (corresponding roughly to the circumference). Usual sizes are 28, 35, 37, 39 and 41 Fr; many designs offer only one or two sizes.

Insertion

DLTs are more difficult to insert than conventional tubes due to their bulk and double angulations: left-sided are somewhat easier to insert than right-sided tubes. Also, because the left upper lobe bronchus is not as close to the carina as on the right, depth of insertion is far less critical. Wherever possible the largest possible left-sided DLT is selected and the cuffs are checked for leaks. Left-sided tubes are appropriate for most procedures except for left pneumonectomy. The device is held so that the bronchial tube curves forwards, resembling the Magill curve of a tracheal tube. It is inserted under direct laryngoscopy and once through the larynx it is rotated counterclockwise so that the tip enters the left main bronchus. Carinal hooks, if present, may snag on the glottis but should ultimately rest on the carina, confirming that full insertion has been achieved. The two connectors are then joined on to those on a twin tube adapter (catheter mount), which links the tubes to the breathing system. The tracheal cuff is then inflated and manual ventilation is commenced via both lumens. This should produce visible equal bilateral chest movement which should be confirmed by auscultation. The tracheal adapter on the catheter mount is then occluded with a hose clamp so that all ventilation is directed down the bronchial lumen. The left bronchial cuff is then slowly inflated whilst auscultating the right lung until a seal is achieved, at which point gas entry to the right lung stops. The tracheal lumen is then released so that the tube can be used as intended.

Insertion of right-sided tubes involves clockwise rotation of the DLT and similar checking of left lung inflation with the added proviso that gas entry should also be confirmed over the right upper zone when the bronchial cuff is inflated.

Both versions may be temporarily reshaped for laryngeal insertion by the insertion of a stiff stylet into the bronchial lumen. This must be removed once the bronchial section has passed through the larynx.

Visual confirmation with a fibreoptic scope is the most accurate method of determining the true position of these tubes and in many centres is now standard procedure. The bronchial cuff should be visible (when the fibreoptic scope is advanced beyond the end of the tracheal lumen) at or just beyond the carina, and with right-sided tubes the origin of the right upper lobe bronchus must be confirmed to lie adjacent to the side hole of the bronchial tube. Large bronchoscopes will not pass into a DLT and a fibrescope of <5 mm diameter is needed. An alternative intubation technique passes the fibrescope through the bronchial lumen and into the correct bronchus before railroading the DLT into place.

If the patient poses a difficult laryngoscopy placement of a DLT may be impossible. Use of DLTs by infrequent users may lead to confusion due to their complexity. In addition, like all tracheal tubes their position may change with patient movement. As DLTs are usually placed when the patient is supine, but surgery is performed in the lateral position (with the table 'broken' to open the rib cage), it is vital to ensure that the tube still allows isolation and ventilation of the lungs as intended after the patient is re-positioned for surgery. Finally, the large size of the DLT and the distance it is inserted makes laryngeal and tracheal injury more likely than conventional tracheal tubes.

Tubes to assist intubation

Tracheal tubes have been designed to assist intubation both during conventional and unconventional intubations.

The *Mallinckrodt Endotrol* tracheal tube (Covidien, Ireland) has a control wire connected to the distal tip that runs in the wall of the tube on the inner aspect of the curvature. Traction on this wire increases the curvature at the tip of the tube to facilitate intubation. This device is no longer available in Europe.

In another design *ETView tracheal tube* (ETView Ltd, Misgav, Israel) a video camera is incorporated in the tip of the tracheal tube allowing visual confirmation of the passage of the tube tip during insertion and constant observation of the tracheal lumen after placement.

The *Intubating LMA tracheal tube* (Intavent Direct, Maidenhead, UK) (Fig. 6.15), which is produced both in reusable and single-use versions, is a valuable tube for use in difficult intubations. It is a reinforced silicone tube with a soft bullet-shaped silicone tip. This tip dramatically reduces the likelihood of impingement on the arytenoids (a common problem with conventional tracheal tubes) and largely eliminates 'hold-up' of the tracheal tube during insertion. This reduces failure, speeds insertion and minimizes the likelihood of laryngeal trauma. Although it was designed for use with the ILMA its performance characteristics make it also suitable for use with catheter exchange techniques and with a flexible endoscope.

Flexible endoscopic intubation does not allow visualization of the glottis as the tracheal tube is advanced. Many rigid optical laryngoscopes and various other devices which guide the tube into the larynx similarly will not permit visualization or easy realignment of tube passage. When conventional tracheal tubes are used with such devices there is a high incidence of difficult insertion and impaction. Many manufacturers now produce their own branded designs aimed at eliminating *hold up* of tracheal tubes passed using systems that do not permit concomitant direct visualization of the glottis as the tube is advanced. None should be assumed to outperform other designs or more particularly to function in other scenarios.

'Hold up', where the tip of the bevel of a conventional tube is not placed centrally in the glottis and snags usually on the right arytenoid or cord, is a more common problem wherever intubation is carried out without a direct view of the larynx. 'Hold up' is thus also possible whenever the tube is advanced unseen over an introducer of any type, be it a gum elastic bougie or a flexible endoscope. As the tube is advanced, the bevel tip is rarely central and impinges on the right cord, arytenoid or posterior rim of the glottis. The conventional solution is to rotate the tube through 90° or more. (See also above: Tracheal Tubes, Size.)

Tubes with additional ports/lumens

Tracheal tubes may have an additional narrow-bore tube fitted to the wall for sampling, pressure monitoring or drug instillation. Sheridan (Hudson RCI, USA) produce a full range of sizes including uncuffed paediatric tubes with an extra lumen ending at the tracheal tip for airway gas sampling or drug instillation.

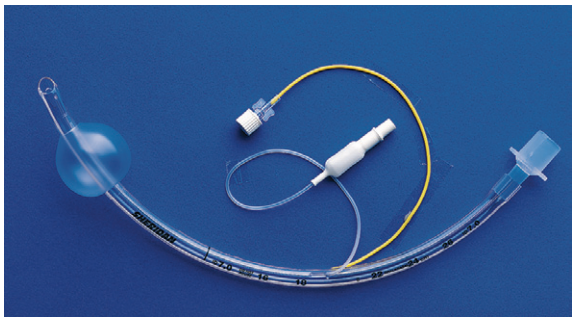


Figure 6.53 Sheridan LITA (Laryngotracheal Instillation of Topical Anaesthetic) tube. A black horizontal line indicates uppermost opening.
Photo courtesy of Hudson RCI.

LITA tube

The extra lumen in this tube has ten small side holes, eight above and two below the cuff for the 'Laryngotracheal Instillation of Topical Anaesthetic'. There is evidence demonstrating better tolerance of intubation in sedated patients[73] and conflicting evidence regarding suppression of response to extubation[73,74] (Fig. 6.53).

Tubes for non-operative jet ventilation (e.g. intensive care) Two additional smaller tubes may be incorporated into the tracheal tube lumen to allow jet ventilation and simultaneous monitoring of tracheal pressure. The much wider main lumen of the tracheal tube then permits entrainment of gasses and passive exhalation at low pressures. Such tubes are available in cuffed and uncuffed versions depending on the need for airway protection and control of entrained gasses.

Developments in tracheal tube and cuff technology for intensive care

The humble tracheal tube cuff is perhaps a neglected item of medical technology. Awareness of both chronic complications of prolonged intubation (subglottic/tracheal stenosis) and incomplete protection (micro-aspiration and ventilator associated pneumonia) have led to renewed interest in improving its performance characteristics.

Cuffs made of silicone or polyurethane are available that eliminate microfolds without the need for high intracuff pressures. Use of continuous electronic cuff pressure measurement and 'servo' technology enables a user-defined intra-cuff pressure to be maintained continuously, even during patient movements: a common cause of cuff leaks (e.g. Venner PneuX P.Y., Intavent Direct, Maidenhead, UK). A tapered (conical) shape to the inflatable tracheal tube cuff may be effective in preventing it

developing folds and microchannels across the full contact surface with the tracheal wall (Mallinckrodt TaperGuard tracheal tubes, Covidien, Ireland).

Addition of appropriately designed small channels opening just above the cuff enables both subglottic suction and lavage.

Finally antimicrobial agents (such as silver nitrate or elutable agents) may be incorporated onto the surface of into the fabric of the tracheal tube.[75,76]

While the benefit of these innovations is largely unproven, each (or perhaps combinations of them) may in the future lead to distinctly different tracheal tubes designed for prolonged use, such as in intensive care or when special protection is required.

SUBGLOTTIC DEVICES

Tracheostomy tubes

The distance between the tracheal stoma and the carina is short and variable. Tubes placed in the trachea via a tracheostomy are, therefore, designed to be non-bevelled, short in length and with the cuff bonded closer to the tip of the tube to prevent accidental bronchial intubation. They may be evenly curved along the length or angulated more sharply to describe a near right angle bend.

There are many varieties of tracheostomy tubes (Fig. 6.54) and only some of the features are detailed below. At least one manufacturer, Rüsch, will tailor single tubes for individual requirements. For intensive care and theatre use simple tubes are appropriate. It has become conventional when tracheostomies are used outside critical care areas and in the community to insert a tube with a liner. A liner is a usually plastic tube that lies within the main tube: as it reduces the internal diameter of the tracheostomy it is not suitable for patients with very limited respiratory reserve. The advantage of a liner is that it can be easily removed for claning of secretions without having to remove the tracheostomy tube itself. When a liner is not used humidification and nursing care must be of a high standard to prevent the build up of secretions.

- *Cuffed/uncuffed*. In children and in adults, where positive-pressure ventilation is not required and the risk of aspiration is not increased, a plain uncuffed tube may be used to maintain the stoma. These may be suitable for long-term use where there is no risk of aspiration (i.e. good bulbar function and good cough).
- *Fenestrated/non-fenestrated*. Fenestration of the tube (a hole that lies below the larynx when the tracheostomy is in place) allows exhalation through the glottis for the purpose of vocalization. Use of a non-fenestrated inner tube in a cuffed fenestrated tube permits positive-pressure ventilation where required. Care must be taken to ensure the fenestrated part of the tube lies within the trachea and not in the pre-tracheal tissues.
- *Long stem/flanged*. In patients with significant swelling or pathology about the head and neck (e.g. patients with burns) ordinary tracheostomy tubes are often too short to reach the trachea. Many manufacturers produce long-stemmed tubes, usually with an adjustable flange (Fig. 6.54) and sometimes in conjunction with an armoured tube wall design, allowing the tube to bend where required rather than having a fixed preformed curve.
- *Foam cuff, Lanz device, Brandt device*. These are available as with normal tracheal tubes as strategies to deal with the problems of cuff seal pressure within the trachea (see above, Tracheal tube cuffs).
- *Above cuff suction/vocalization*. A soft catheter ending above the tracheal cuff may be bonded to the tube to permit suction removal of secretions that tend to pool in the larynx and which may contribute to ventilator associated pneumonias. A similar design can be used to insufflate air to permit vocalization even during positive pressure ventilation.



Figure 6.54 Just a few of an immense range of tracheostomy tubes (see text). The bottom left hand device is a tracheostomy tube with an adjustable flange allowing correct positioning of the distal portion of the tube in patients with bigger or oedematous necks.

- *Distal pressure monitoring*. A narrow bore tube ending distal to the cuff allows direct measurement of tracheal pressures.
- *Speaking valves*. The tube may be capped with a one-way valve that allows gas to enter on inspiration but prevents gas passage on expiration. Provided the cuff is deflated expiratory gas is redirect through the larynx which allows vocalization. If the cuff is inflated the valve *must* be removed.

Percutaneous tracheostomy kits

Percutaneous tracheostomy has become a routine procedure on intensive care units in the last two decades. The popularization of a percutaneous approach, as opposed to an open surgical technique, dates from Ciaglia's description of a guide wire and serial dilator method in 1985.[77] Briefly the technique is as follows: the trachea is punctured with a needle through which a Seldinger-type guide wire is placed. A thin stiffener is inserted over the wire, then a series of increasingly large dilators are used to enlarge the stoma. Once the largest dilator has been passed, it is removed and the tracheostomy tube carried on a smaller dilator is inserted over the wire. The Ciaglia Blue Rhino (Cook Medical, USA) (Fig. 6.55) has since superseded serial dilation with a simpler one-step tapered dilator and this modification has been adopted by other manufacturers.

Percutaneous dilatational tracheostomy (PDT) has become routine in ITUs in European and other industrialized nations. Complications include bleeding, misplacement, loss of the airway and needle or blunt trauma to adjacent structures. Numerous variations on PDT have been devised aiming to address these issues (see below). PDT appears to carry a perioperative complication rate comparable or superior to standard tracheostomy, but there are still differing opinions regarding the overall advantages or otherwise, beyond the clear organizational issues in intensive care.[78–80] For guide-wire-based techniques, bronchoscopy has become an accepted standard of care, to ensure that the tracheal puncture is central and well-positioned and that the posterior tracheal wall is not breached. Management of the upper airway during the PDT may be via the original tracheal tube, a new tracheal tube or via a supraglottic airway: several solutions have been advocated. Use of bronchoscopic guidance should reduce complications and eliminate paratracheal placements. Perioperative bleeding during percutaneous techniques is usually tamponaded by the tracheostomy tube but can be troublesome and occasionally major. PDT is essentially an elective procedure with reported operative times of 5 to 15 minutes.

Other approaches to percutaneous tracheostomy are less widely used, but include the following:

- The Rapitrac uses sharp forceps that are held open to admit the tracheostomy tube, but these are prone to damaging the tube cuff and tracheal wall; it has been largely abandoned.
- By comparison, the Portex guide wire dilating forceps (GWDF) technique (Fig. 6.56) described by Griggs in 1990[81] uses blunt curved forceps inserted over a guide wire to dilate the stoma, which are then removed. Griggs' technique may be applicable in emergencies as it is quicker to perform than serial dilation but may carry a higher complication rate.[82]
- The Percutwist from Rusch, uses a screw-type dilator to follow the guide wire. Limited evaluation of the technique is available.[83]

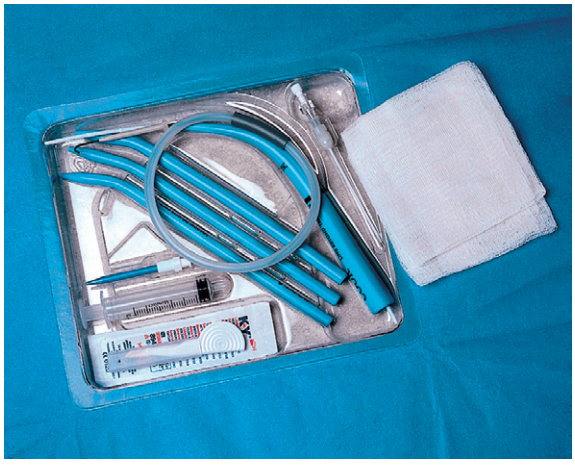


Figure 6.55 Cook Blue Rhino single step percutaneous dilatational tracheostomy kit.
Photo courtesy of Cook UK.

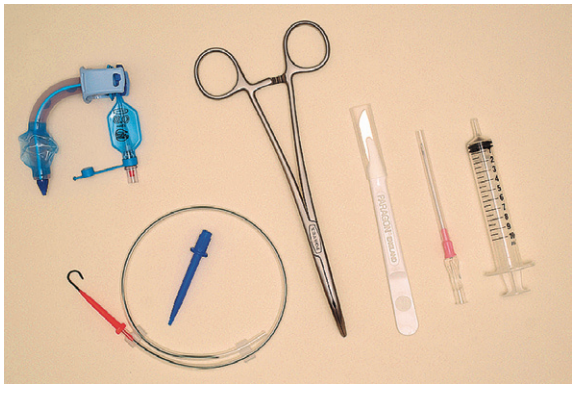


Figure 6.56 Griggs Guide Wire Dilating Forceps percutaneous tracheostomy set. The forceps are recessed on the inner surface of the prongs along the distal curve to follow the guide wire.

- Cook have recently introduced a balloon dilation technique (Ciaglia Blue Dolphin) where dilatation is achieved by a balloon advanced over the guide wire.[84] This is then inflated with saline up to a pressure of 11 atmospheres to dilate the tissues, after which the tracheostomy tube is inserted. The manufacturers claim a reduction in the risks of tracheal ring and posterior tracheal wall injuries as serial dilators are not forced in and out.
- A differing approach from Mallinckrodt is the translaryngeal technique of Fantoni,[85] which is unique in that it involves creating a stoma from inside to outside. A specially designed dilator tipped tracheostomy tube is inserted through the larynx and pulled out through the tracheal wall. Its use is not widespread.

Devices for cricothyroidotomy

Smaller diameter cannulae are simpler, safer and quicker to insert percutaneously into the trachea than formal tracheostomy tubes, particularly if the cricothyroid membrane is used for access. Ventilation (as opposed to oxygenation), however, is only possible if high-pressure gas is jetted through the device, with exhalation occurring passively through the larynx. Transtracheal jet ventilation (TTJV) is therefore the 'bottom line' in algorithms for the management of failed ventilation and is used electively in other scenarios such as for laryngeal surgery or in the management of the difficult airway, where it may provide ventilation during fibreoptic intubation or tracheostomy under general anaesthesia.[86]

A variety of larger-bore devices for emergency tracheal access, some utilizing guide-wire techniques for insertion, are made by several manufacturers. They tend to be an unhappy compromise between allowing low pressure 'to and fro' ventilation and ease of insertion, with no proven advantages over more formal percutaneous tracheostomy. They may have a role for 'out of hospital' use but little is known of the long-term effects of attempting to put larger devices through the cricothyroid membrane.[87]

Manujet 'injector' (VBM Medizintechnik)

This is a modern version of the manually operated Sanders high-pressure gas injector for 'jet ventilation' (Fig. 6.57). By squeezing the trigger, oxygen from a standard 4 bar Schrader socket can be intermittently discharged through a suitable cannula or catheter. The Manujet incorporates a pressure regulator that allows jetting pressure to be varied down to below 0.5 bar for use in infants.

As the Manujet may be supplied with a standard Schrader probe (for connection to wall oxygen) or a 'mini-Schrader' (small diameter quick release probe for connection to an anaesthetic machine high-pressure source) it is essential that the correct configuration is chosen.

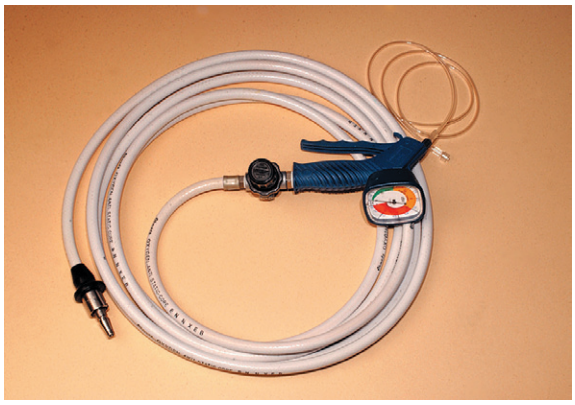


Figure 6.57 Manujet, variable pressure manual injector, VBM Medizintechnik, Germany.

Ravussin catheter

The Ravussin Jet Ventilation Catheter (VBM Medizintechnik, Salz, Germany) is probably the device of choice for use as a cricothyroid cannula and is developed from the intravenous cannulae which were originally used and advocated for TTJV.[88,89] It is a 'cannula over needle' device available in 13G for adults and smaller 14G and 18G versions for children and babies. The kink-resistant catheter is made of Teflon and is curved with an angled flange on the connector so that it points axially down the trachea when it is fully inserted. The proximal end has a Luer-Lock for connection to a high-pressure gas supply housed within a 15 mm ISO male connector which allows *very temporary* low-pressure oxygen insufflation from a conventional anaesthetic system. The distal tip of the catheter has three small lateral holes to centre the device in the trachea during jetting and prevent 'whipping'. Peak airway pressures generated at the carina during low-frequency TTJV at 4 bar are of the order of 5–13 mm Hg.[90]

The device (Fig. 6.58) with a syringe attached, is inserted in a caudal direction through the cricothyroid membrane, with aspiration of air confirming intratracheal needle placement. The syringe assembly is held still whilst the catheter is advanced off the cannula. The syringe is then reattached to aspirate air and confirm catheter placement. If care is not taken at insertion it is easy to end with the catheter pointing upwards and through the larynx: air will be easily aspirated with a normal capnography trace but jet ventilation will be ineffective. Partly for this reason a test inflation is advocated in elective use. The Ravussin cannula is well tolerated by awake patients and may be left in place prior to or after anaesthesia without inconvenience.

Several studies have shown that for emergency airway rescue, assembly of a suitable system from disparate items during an emergency is not practicable, taking too long and leading to ineffective systems. Intravenous cannulae

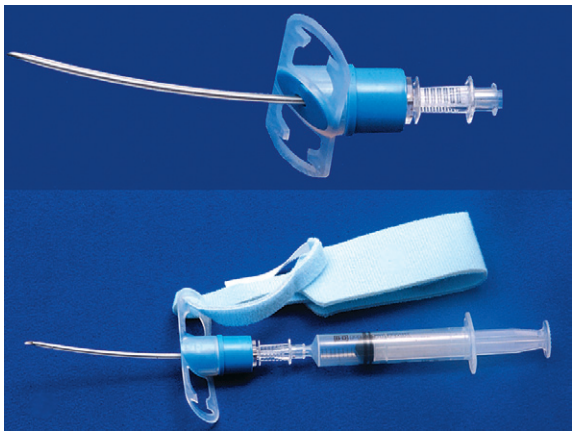


Figure 6.58 Ravussin transtracheal jet ventilation catheter.

are not designed for this role: they are difficult to fix at the neck and almost inevitably kink making their continued use impossible. A device such as the Ravussin together with a high-pressure oxygen source and a means of delivery (e.g. Manujet system) should be available in all areas where anaesthetics are given and where unconscious patients may be admitted.

Melker cricothyroidotomy kits

These are Seldinger type kits supplied in a 6.0 mm ID uncuffed design and a more recently introduced 5.0 mm ID cuffed design from Cook Medical. The kit comprises a needle, a stab scalpel, a guide wire, a dilator-introducer and the 9.0 cm long tube. The needle is inserted and a semi-rigid guide wire is passed through it. A 2–5 mm 'stab incision' is made around the guide wire and the cricothyroid tube mounted on a dilator-introducer is *advanced as a single unit* along the wire and into place. Once the introducer is removed from the tube and the cuff (where present) is inflated, the tube is ready to use. Lateral flanges can be used to secure the tube to the neck with tape or stitches. Evaluations show rapid deployment, high success rates and high levels of user satisfaction.

Quicktrach

This is a cannula over needle device from VBM Medizintechnik, available now in both cuffed and uncuffed versions. There are three sizes for adults, children and neonates (4.0 mm, 2.0 mm and 1.5 mm ID respectively). The device is relatively bulky and also, as it is considerably shorter than the Melker device, there is risk that in oedematous or obese patients it will be too short to be adequately inserted into the tracheal lumen. The use of a large bore cannula over needle design offers speed of insertion but risks significant trauma and difficulty during placement.

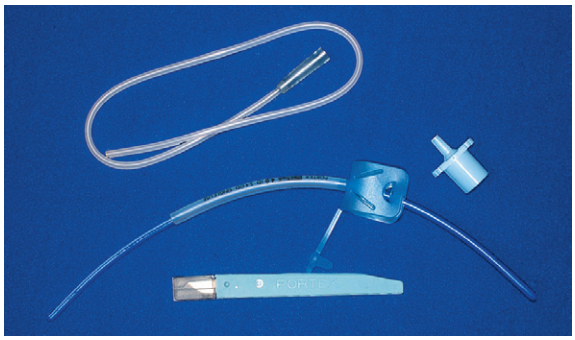


Figure 6.59 Portex Mini-Trach kit, Smiths Medical.

The Mini-Trach

This is a simple 4.0 mm ID uncuffed tracheostomy tube (Fig. 6.59), which is now also available in a Seldinger technique kit. The Portex kit was originally designed for tracheal toilet and being an uncuffed small bore tube is not ideal for 'to and fro' ventilation. Rather large 'steps' between cannula and catheter make it potentially difficult to insert and therefore not ideal for emergency use.

Portex Cricothyroidotomy Kit (PCK)

This relatively new cricothyroidotomy device, originally designed for military use, deploys a pre-assembled 6.0 mm ID cuffed tracheostomy tube. Its novelty lies in the fact that it uses a Veress needle to indicate passage into the air-filled trachea, thereby increasing ease of use and limiting, in theory, risk of damage to the posterior tracheal structures.

LARYNGOSCOPES

Laryngoscopes are designed to allow the larynx to be seen and thus to enable tracheal intubation. This is usually easily achieved, but is occasionally very difficult. They may be considered under two broad categories:

1. *Direct line of sight devices – rigid lighted retractors*, such as the Macintosh laryngoscope, are reliant on retracting tissues to create an uninterrupted sight line between the operator and the objective. Fibreoptics may be used in the light source of these types. This is the traditional (and still routine) technique used for the vast majority of tracheal intubations. It is termed *'direct laryngoscopy'*.
2. *Indirect line of sight devices – optical laryngoscopes*, where a fibreoptic bundle, and/or series of lenses, prisms or mirrors, or now even a minature camera, transmits an image to the user from the distal end of the device. This enables the observer to effectively view wherever the device tip is pointed and thus to 'see around corners', meaning that the 'line of sight'

requirement of direct laryngoscopy is obviated. The general trend in this group is for increasing use of digital video and screen miniaturization technology for image transmission (*videolaryngoscopy*) – doing away with the need for fragile expensive, fibreoptics. *Optical laryngoscopes* can be subdivided into the following:

a. *Rigid optical laryngoscopes*, where the image conveying system is encased in a rigid structure. This type of device can manipulate and displace soft tissues, acting as a retractor; however, the predetermined and fixed shape of these devices limits their applicability. Accepted wisdom has it that such instruments require less dexterity and expertise to use than their flexible counterpart. There seems to be a new product daily in this class at the moment; none are of proven value in general use, nor do they dominate in any particular application.

b. *Flexible endoscopes*. In the flexible fibreoptic endoscope (fibrescope) the viewing bundle plus light transmission bundles and an optional instrument channel are wrapped in a flexible casing. The instrument can thus be made to follow anatomical spaces and will bend as necessary to negotiate almost any route. Digital camera technology can also be used to convey the image from the tip, and a new single-use device – the Ambu aScope (see below) – contains no fibreoptic bundles, but is identical in function to a fibrescope. The term *flexible endoscope* is therefore more accurate in general for these devices, but it will be a while, no doubt, before the *argot* of anaesthesia reflects the new times. Flexible endoscopes cannot retract tissues and require pre-existing space (e.g. an awake breathing patient, or if need be a purpose-built artificial airway) in order to provide a view.

History

Visualization of the vocal cords for intubation was popularized by Sir Robert Macintosh and Sir Ivan Magill in the early 1940s. It was during the insertion of a Boyle–Davis gag that Macintosh conceived the idea of his laryngoscope, which is still the most popular design in use today and has spawned a wide variety of modifications. It consists of a blade that elevates the lower jaw and tongue, a light source towards the tip of the blade to illuminate the larynx and a handle to apply force to the blade. The handle also contains the power supply (battery) for the light source. The light comes on when the blade, which is hinged on the handle, is opened to the right-angle position.

Macintosh designed a slightly curved blade (Fig. 6.60) with a small bulbous tip that was designed to be inserted



Figure 6.60 Macintosh laryngocope with three sizes of interchangeable handle.
Image courtesy of Timesco of London, UK.

anterior to the base of the epiglottis in an adult. The child and infant blades were not designed by him and he condemned them as being anatomically wrong and unnecessary. Most blades for infants and children and some of those for adults tend to be either straight or with a small shallow curve at the tip only. These are designed to be inserted deeper into the pharynx and posterior to the epiglottis and hence the blades are correspondingly longer.

In UK practice, the term 'laryngoscope', when not further described, is still largely synonymous with the rigid retractor type and, more specifically, the Macintosh designed laryngoscope. The term 'difficult laryngoscopy', therefore, is largely used to describe any situation where the best view of the larynx with these default device is suboptimal; often defined by inability to see any part of the glottis. Given the variety of devices now routinely available, and the different laryngoscopy techniques required with the various retractor and optical type laryngoscopes, the terms 'difficult laryngoscopy' and 'difficult intubation' are evolving and should always be further defined to describe the circumstances both in terms of views achieved and equipment used.

Retractor type laryngoscopes

Fig. 6.61 shows some of the wide variety of blades currently available. The choice of blade for routine use is probably largely a matter of personal preference. It must be borne in mind that the technique for laryngoscopy is different for the various designs of blade and differing designs may offer better views of the larynx in a given patient. Logically, therefore, a variety of blades should be

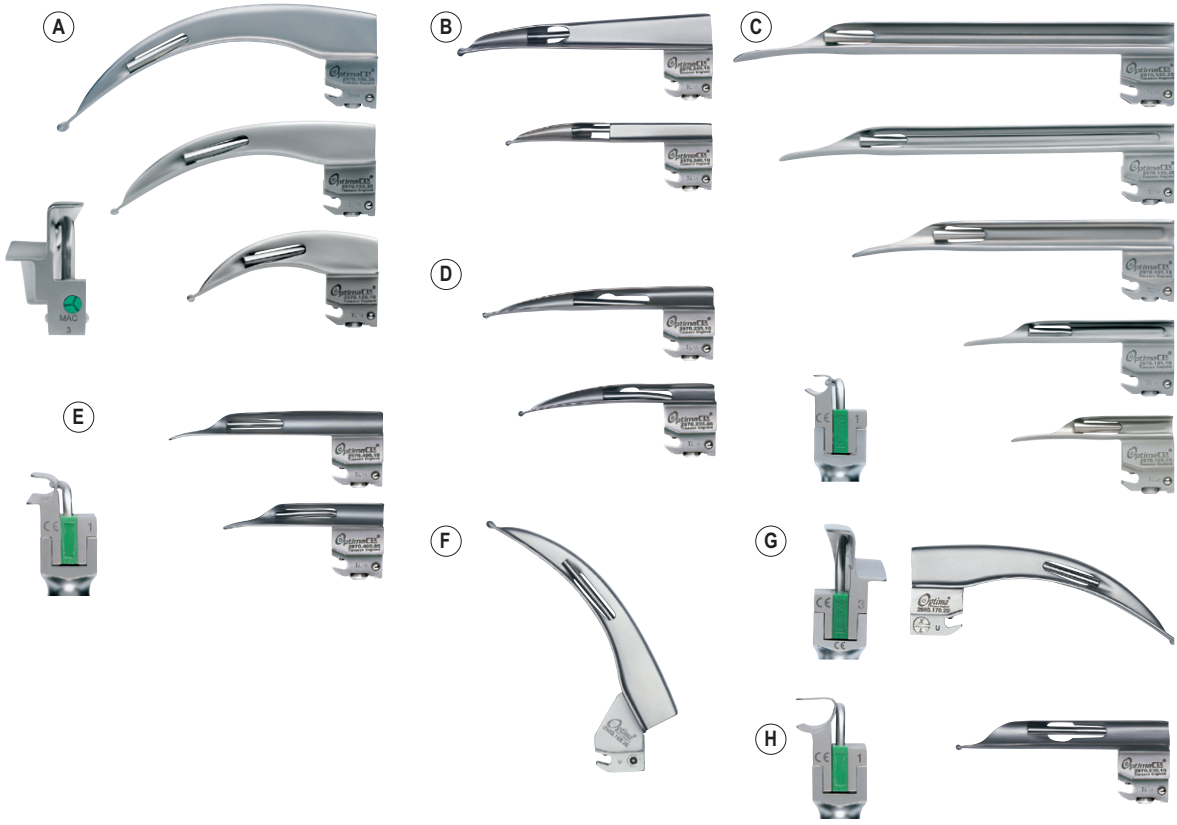


Figure 6.61 Laryngoscope blades. **A.** Macintosh pattern sizes: large adult, adult, infant. **B.** Seward pattern sizes: child, infant. **C.** Miller pattern sizes: large adult, adult, child, infant, neonate. **D.** Robertshaw pattern sizes: infant, neonate. **E.** Wisconsin pattern sizes: infant, neonate. **F.** Polio pattern sizes: adult. **G.** Macintosh left-hand pattern size: adult. **H.** Oxford pattern size: infant. *Image courtesy of Timesco of London, UK.*

available wherever anaesthesia is practised and all anaesthetists should develop skills with more than one type of blade. Most blades are detachable from the handle for ease of cleaning and change of blade size where appropriate. The 'hook on' connection for the blade, which allows easy detachment, is very convenient and was developed by Welch Alleyn Ltd in the early 1950s.

Two standards – ISO 7376/3 (green system) and ISO 7376/1 (red system) – are in existence that allow blades from different manufacturers to be interchangeable, but they have not been universally adopted. In both the bulb is housed within the handle and light is transmitted through an optical 'bundle' to the tip of the blade. Their difference lies in the dimensions of the hinges and the relative positions of the light sources.

Prisms and mirrors are sometimes added to these laryngoscope blades to overcome the principal shortcoming of this class of device, namely that the operator's eye and patient's larynx must be aligned with no interposing tissues. So far, such modifications have not proven popular or lasting.

Features of modern laryngoscopes

Fig. 6.62 shows a typical instrument with a hook on type Macintosh blade. Some of its specific features are highlighted below:

- detachable blade for interchangeable blade designs and ease of cleaning and sterilization
- light source sited within the handle. Much brighter xenon gas filled bulbs are used to compensate for light loss during transmission
- light projection via a shaped bundle of glass fibres. The bundle may be manufactured as an integral part of the blade, or may be detachable so that should it become damaged or opaque it may be replaced separately. Fibreoptic bundles are prone to degradation resulting in poor illumination[51] and difficult laryngoscopy.

Single-use and reusable laryngoscope blades

Recently much has been made of the failure of standard sterilization techniques to eliminate all proteinaceous

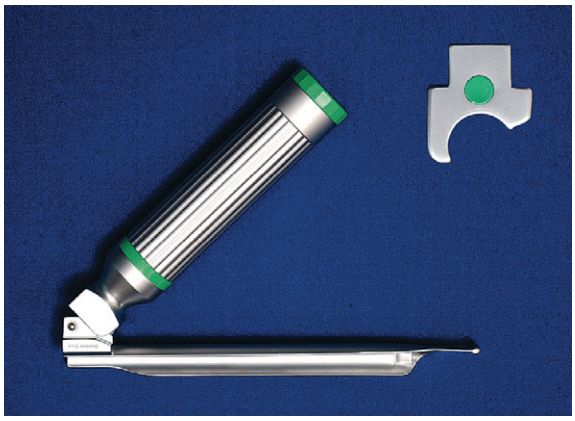


Figure 6.64 The Henderson blade attached to a laryngoscope handle (here in 'ISO Green system' fitting). Inset shows the blade from the rear to demonstrate the profile in cross-section.

and associated soft tissues into the elastic and distensible area of the floor of the mouth (mandibular space), providing an uninterrupted sight line through to the larynx. Poor views of the larynx can be predicted from this model where there is:

- inadequate craniocervical movement or jaw opening
- relative reduction in the distensible area below the floor of mouth (e.g. a small receding mandible, scarring or distorted anatomy, for instance following head and neck surgery or radiotherapy)
- increased volume or decreased compressibility of tissues needing to be retracted (e.g. swelling or malignancy of tongue)
- lesions otherwise obstructing the line of sight or precluding ideal laryngoscope positioning (e.g. at the vallecula).

Such an analysis will detect many, but not all, difficulties.

The handle of the laryngoscope is used to *lift* (i.e. force is applied in the direction of the handle) rather than using the handle as a lever. Curved blades are designed for the tip to be inserted into the vallecula with the standard Macintosh blade being inserted to the right of the tongue, while displacing it to the left side, whereas the straight blade may be inserted posterior to the epiglottis and is particularly useful for small children and adults with a large floppy epiglottis.

Different laryngoscope blades require different techniques for viewing the larynx, which must be learnt and used to maximize utility of that device. For example, the Henderson blade (Karl Storz, Germany), a modification of the Miller blade, is a long straight blade with a 'C'-shaped cross-section (Fig. 6.64) and is intended to be inserted to the right of the tongue behind the molar teeth with the head turned aside: a 'retro-molar' approach.

A poor view obtained with one laryngoscope design does not necessarily predict a poor view with a different design: ENT surgeons using a cone-shaped straight operating laryngoscope may, often to the annoyance of the anaesthetist who has struggled to intubate the patient, proceed to operate on the very glottis that the anaesthetist could not visualize.

There are numerous other laryngoscope blades beyond the designs already referred to: a few are briefly described here.

Polio laryngoscope

This variation (Fig. 6.61F) comprises a normal Macintosh blade, but the angle between the blade and the handle is increased from 90° to 135°. This therefore enables the laryngoscope blade to be more easily inserted into the mouth in patients with abnormal anatomy, e.g. limited neck extension, large breasts, or those in unusual situations such as patients in a cabinet ventilator for polio: the circumstances for which it was developed. Another device, the *Patil–Syracuse* handle, allows the handle and blade to be locked together at a variety of angles.

McCoy blade

This is based on a standard Macintosh blade modified by the insertion of a hinge to give an adjustable tip that is operated by a lever on the handle (Fig. 6.65). The blade is inserted in the normal way, and if the view is obscured, the tip can be flexed so that it further elevates the vallecula and epiglottis.[97] It has been shown to improve some 'awkward' laryngeal views: for example the view of the larynx in 70% of patients with Cormack and Lehane grade 2 or 3 views, when restrained by manual in line stabilization, can be improved.[98] However, it is likely to have little effect on genuinely difficult (grade 4) laryngoscopy.[99] Its bulk makes it rather cumbersome, perhaps contributing to making easy intubations slightly less easy.[100, 101] As a result is it not recommended for routine use.

Flexiblade

The Flexiblade takes the McCoy concept a step further. The laryngoscope blade is made of multiple sections joined together, which in its 'unflexed' position resembles a Macintosh blade. Deployment of a lever after insertion of the blade flexes the whole length of the blade drawing the tip upwards in a similar manner to the McCoy blade. There are limited, but mostly positive, evaluations.

Rigid optical laryngoscopes

Fibreoptic technology dates back to the 1950s and is described in greater detail below under flexible fibreoptic laryngoscopes. In anaesthesia, flexible fibreoptic endoscopes predate the rigid devices described in this section. Nonetheless, rigid fibreoptic laryngoscopes (e.g. Bullard and Upsherscope) have been available for several decades

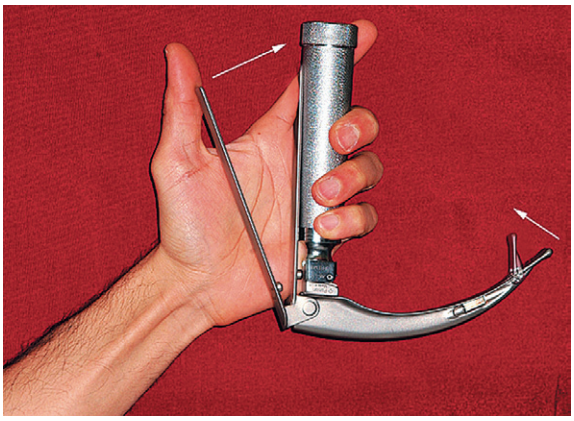


Figure 6.65 The McCoy laryngoscope with the lever deployed to show flexion of the tip.

and have long been advocated for use in difficult airway management by enthusiasts. In recent years a reduction in the costs and size of the components, allied with improvements in video technology, has led to a profusion of new entrants to the market. Image transmission in this class of device may be by fibreoptics, lenses/prisms/mirrors, or increasingly, electronically from a distally mounted miniaturized video camera (i.e. videolaryngoscopy).

Rigid optical laryngoscopes offer the following potential benefits:

- the ability to convey an image removes the need for a direct line of sight, making intubation theoretically possible when direct laryngoscopy fails. (The term *indirect laryngoscopy* is tempting, but for many will always carry connotations of suave Ear Nose and Throat specialists with head mounted and hand-held mirrors)
- the ability to retract tissues and make a space to look into or beyond (which flexible endoscopes can not do)
- the ability to gain a view of the larynx with less mouth opening
- reduced force and distortion of tissues leading to:
  - reduced cardiovascular stimulation
  - the ability to use in awake patients after topical anaesthesia
- the facility to add video projection of the viewed image.

The addition of video technology to laryngoscopy has several benefits:[102]

- technical – a larger, brighter, higher-resolution image
- procedural – allowing instruction and demonstration to multiple viewers
- educational – static images or live video feed can be captured and reviewed
- documentation – recording of pre-/intra-/post-procedural images
- research – allows easier comparison between patients, procedures and operators
- user comfort – physical comfort from improved posture.

However, these benefits of the rigid optical laryngoscopes are all only of value if the laryngoscope design reliably facilitates and achieves tracheal intubation.

Problems with rigid optical laryngoscopes stem from:

- large numbers of devices with limited proof of efficacy
- potential difficulty in achieving tracheal intubation despite a good view of the larynx, a common finding which may be caused by a mismatch between where the device views as opposed to where it steers a tracheal tube. This may be overcome by a number of manoeuvres which vary between laryngoscope types
- increased difficulty and time to achieve intubation of easy cases
- necessity to use stylets and specialized tracheal tubes to achieve intubation
- trauma to unsighted areas in the airway during passage of tracheal tubes (and stylets)
- increased cost, a very major issue with many devices costing in excess of £5000 ($7000), often with disposable components also required. This leads to secondary issues:
  - reluctance of manufacturers to loan devices meaning choice to purchase is rarely based on local experience
  - reluctance of manufacturers to loan devices limiting research opportunities
  - increased cost per patient: some devices may cost >£100 ($150) per use.

While rigid optical laryngoscopes have much potential, at present the majority of devices are of unproven benefit, in both routine and difficult cases. Much of the research that has been performed on these devices is of poor quality[103] with extrapolation of results from easy patients to difficult patients, use of intermediate endpoints and an emphasis on cohort rather than comparative studies.

The exact role for these devices is as yet undetermined, given: their expense which precludes them being the default laryngoscope for routine use, their need for specific training over and above that needed for say a Macintosh laryngoscope, and the fact that ultimately they do not preclude the need for flexible endoscopes.[104,105] They are also principally limited to orotracheal intubation.

There are too many devices to describe in detail but a selection of devices is briefly described below, under three broad groups. Note again that many of these devices contain no optical fibres, relying instead on image transmission from a distally placed imaging CCD (camera chip) and illumination by a similarly located LED:

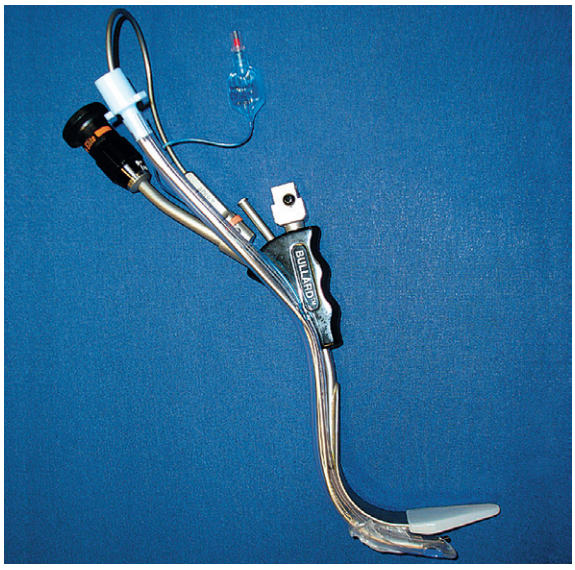


Figure 6.66 The Bullard laryngoscope here with light handle not connected. The tracheal tube is mounted on the integral stylet.

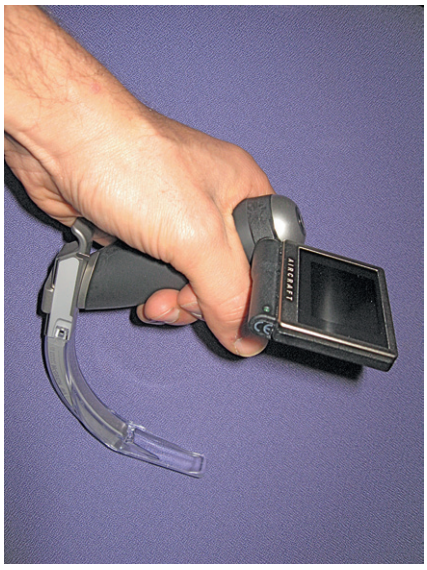


Figure 6.67 Magrath Series 5 video laryngoscope. Aircraft Medical, UK.

- bladed type rigid optical laryngoscopes
- conduit type rigid optical laryngoscopes
- optical stylets.

Bladed rigid optical laryngoscoscopes

Bullard laryngoscope

This device was for many years the 'standard' for this group but has to a large degree now been superseded. It has a broadly curved blade and uses fibreoptics to transmit the image from the tip to the eyepiece. It is designed to elevate the jaw without the need for neck extension and for use in patients with limited mouth opening. A fixed stylet is incorporated in later models to carry the tracheal tube (Fig. 6.66). The WuScope is a further example of this type. This class of devices appears to have enjoyed greater popularity in the USA.[106]

McGrath Series 5

The McGrath (Aircraft Medical, UK) laryngoscope is an ergonomically engineered, bladed, rigid video laryngoscope with an integral pivoting screen and detachable metal CameraStick. A disposable blade made of a stiff clear plastic covers the CameraStick and is curved with a mid-portion angulation (Fig. 6.67). At present there is only an adult size; a ratchet mechanism at the base of the device enables a 3 point retraction of the camera part for use in smaller patients. It is operated by standard batteries, with a good battery life. Case reports have shown its utility for management of several cases of difficult intubation, but large scale comparative studies are still awaited.

GlideScope

The GlideScope video laryngoscopy systems (Verathon Medical, USA) comprise a curved plastic Macintosh-like blade, with a distal tip camera that projects to a remote viewing screen. As with many of these devices there is no use of fibreoptics. It is one of the most popular of the newer intubation aids. Originally the device was fully reusable after sterilization, but recently a newer version, the 'GlideScope Cobalt', has been introduced with a reusable flexible video baton which inserts into a disposable perspex blade (Fig 6.68). There are four different sized handles in the reusable range (two sizes of a different shape for the emergency services version) and five different sized disposable blades in the single-use Cobalt range, hinting at some dissatisfaction with the blade functionality. There are numerous publications on the original GlideScope, although many of them suffered from the weaknesses described previously. Like many bladed rigid optical laryngoscopes there is a danger of 'easy view, difficult intubation', particularly if the manufacturer's instructions are not followed. After various descriptions of use of stylets with varying degrees of bend the manufacturers have introduced specific single-use tubes (GlideRite) and a dedicated stylet (Fig. 6.69). A technique that involves introduction of the styletted tube from the side of the patient's mouth towards the pharynx followed by 90° rotation into the field of view is described; as much of



Figure 6.68 GlideScope Cobalt video laryngoscope system, adult and paediatric handles and a selection of disposable blades and intubation stylet.
Photograph courtesy of Verathon Medical UK, www.verathon.co.uk



Figure 6.69 GlideRite tracheal tube and intubating stylet.
Photograph courtesy of Verathon Medical UK, www.verathon.co.uk

the route of passage is invisible during this technique, care must be taken to avoid soft tissue trauma, which is reported both with the GlideScope and with the McGrath laryngoscope.

C-Mac

The C-Mac (Karl Storz, Tuttlingen, Germany) comprises a reusable metal laryngoscope with blade similar in profile to a Macintosh. The blade houses a camera and LED near its tip (Fig. 6.70A). A second, more curved 'D' blade is also available for the system. The detachable electronics module inserted into the handle to power the unit and transmit the image to a remote screen also permits control of the recording facilities. The system is designed to be usable both as a conventional Macintosh laryngoscope and as a videolaryngoscope. As such it has potential as a teaching tool: the trainee may use it for direct laryngoscopy while the instructor observes or directs progress from the remote monitor. It has been shown to improve training in direct laryngoscopy and use of the camera capability improves visualization of the larynx when compared to use without it.[107,108] It must be noted that the studies referred to here were carried out using a prior device (the Berci DCI Videolaryngoscope) from the same manufacturer (Fig. 6.70B). In this incarnation an externally identical-looking laryngoscope handle bears a short length of image and light-carrying fibreoptics with integral optical connections for the detachable light source and camera – the Direct Camera Interface. Fields of view for the devices are different as is the marked propensity of the older device for fogging. Image quality overall is significantly improved in the C-Mac.

Conduit rigid optical laryngoscopes

Upsherscope

The Upsherscope (Fig. 6.71) is a rigid fibrescope which has many features that are similar to the Bullard scope but instead of a stylet to deploy the tracheal tube there is a integrated channel (conduit) for passage of the tracheal tube, lying posterior to the blade. The optical system also lies within the conduit posterior to the distal part of the blade.

Pentax AWS

The Pentax Airwayscope (Pentax Europe GmbH, Germany) comprises a bulky handle, containing the power source and carrying a multi-positional LCD monitor screen, and a flexible stem terminating in a small unit housing the CCD and LED (Fig. 6.72). A single-use plastic blade is mounted on this and incorporates conduits for both the tracheal tube and a suction catheter. The device is designed to be used in the midline and the tip rotated 90° anteriorly on entering the pharynx. The screen has a fixed 'target symbol' (a cross) on it; positioning the blade such that the laryngeal inlet lies in the middle of the screen is designed to ensure that the tracheal tube passes directly into the trachea. Increasing numbers of (mostly cohort) reports suggest successful use with the Pentax AWS. It is likely to have the same limitations, regarding tube choice, as the Airtraq (see below).

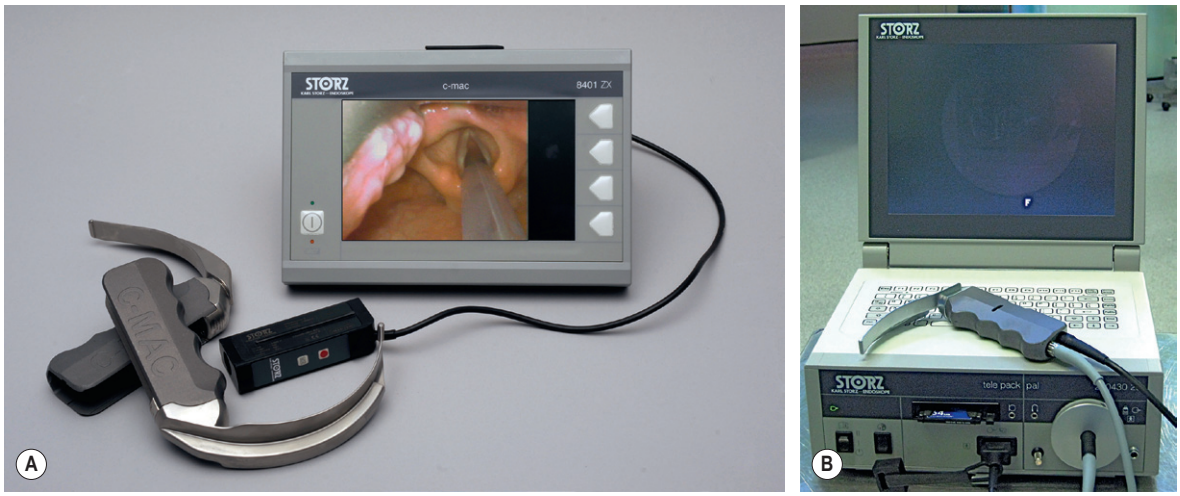


Figure 6.70 **A,** The C-Mac and **B,** the Berci DCI videolaryngoscope systems from Karl Storz, Germany. Note the extreme curvature of the C-Mac 'D' blade (on top), for difficult laryngoscopy, which cannot be used for direct laryngoscopy, see text.

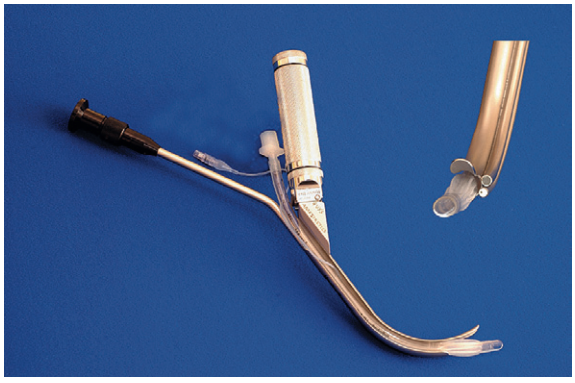


Figure 6.71 The Upsherscope, inset, shows the tip of the device with tracheal tube emerging. The larger light transmission fibreoptic bundle lies above the much smaller coherent bundle for image transmission. The size of the distal viewing lens explains why fibreoptic devices are so easily foiled by secretions.

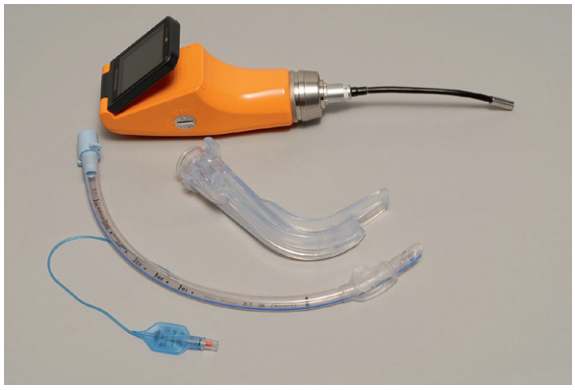


Figure 6.72 Pentax Airwayscope with disposable blade and a tracheal tube.

Airtraq

The Airtraq (Prodol Meditec, Spain) (Fig. 6.73) is a novel device that does not use video or fibreoptic technology, is low cost and is entirely disposable. The device has parallel plastic light and tracheal tube conduits, with the resulting insertion part curved through 90° in its distal third. The user looks into a viewfinder where a series of prisms and mirrors deliver an illuminated image from the tip of the device (Fig. 6.74). The LED light flashes for about 30 s at switch on whilst the anti-fog system is warming up. Insertion is similar to the Pentax AWS, but variations similar to insertion of a Guedel airway are described. A camera may be attached and the image viewed on a remote screen connected by wireless technology. Four sizes of device are available, accommodating tracheal tubes 2.5 to 8.5 mm ID. Further designs attempt to assist nasal intubations and double lumen tracheal tube insertion. It is a commercially successful product, although so far only a few small series show improved success rates at intubation compared with a Macintosh laryngoscope.

All conduited rigid optical laryngoscopes suffer from several limitations. First the conduit increases the bulk of the 'blade' and may increase the required mouth opening compared to simpler bladed devices. Second the intubation is blind until the tracheal tube passes beyond the camera (though the fact the tube is housed within the conduit should limit the risk of trauma). Finally, use of different tracheal tube types and sizes will likely alter the



Figure 6.73 Airtraq optical laryngoscope, Prodol Meditec, Spain.

angle of exit of the tracheal tube from the conduit which may affect intubation success. It has been shown for the Airtraq that use of tracheal tubes that are both preformed and whose external diameter is only a little smaller than the internal dimension of the conduit facilitate intubation; conversely, relatively small and non-preformed tracheal tubes will fail to adopt the shape of the conduit and therefore exit in a more posterior direction leading to a tendency for failure or oesophageal intubation. This effect is likely to similarly affect the Pentax AWS.

Conversely, the conduited devices have the major benefit over those without a conduit that once the larynx is viewed, if the correct tracheal tube is used, passage of the tracheal tube should be simple and reliably achieved.

LMA CTrach

The CTrach (Intavent Direct, Maidenhead, UK) (Fig. 6.75) is a modification of the ILMA, but is appropriately classified as a conduit-type, rigid optical laryngoscope. It comprises a reusable ILMA carrying two fibreoptic bundles that emerge in the bowl behind the epiglottic elevator and transmit light and images to a connector at the proximal end of the airway tube, next to the handle of the device. The epiglottic elevator is modified in design (a hole in the centre of it enables a view beyond) and colour (it is white, to minimize interference with images). After the device is inserted a battery powered, low-definition screen is attached to the connector to display an image. Unlike other conduit devices (but like the ILMA) the design of the CTrach airway tube ensures a variety of sizes of tracheal tubes will exit the device at the correct angle and in the midline. Three sizes of device are available, covering adults of 30 to 100 kg.

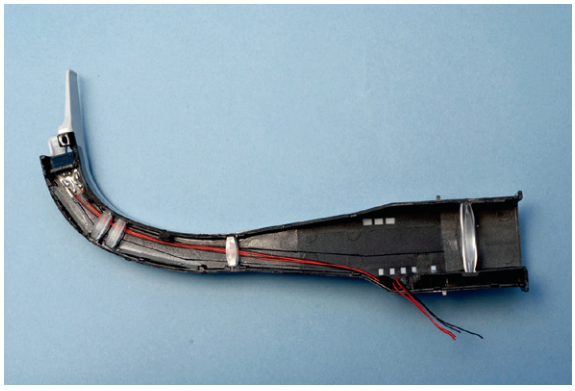


Figure 6.74 Airtraq optical laryngoscope tortured to reveal the light pathway and wires for the LED and anti-fog circuits.

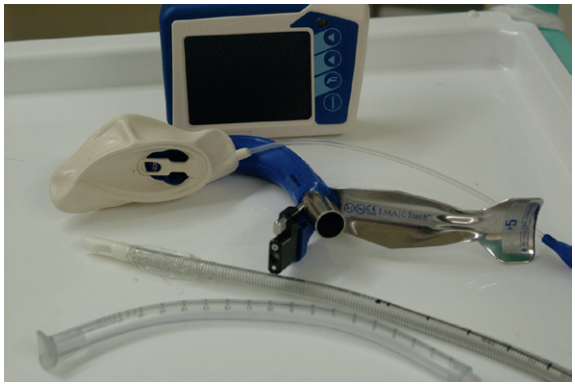


Figure 6.75 LMA CTrach, Intavent Direct, UK.

The CTrach enables high rates of successful intubation, even when the view provided by the fibreoptic system is poor or uninterpretable. Disadvantages include an up to 30% rate for 'no view' (in which case it may be used as a standard ILMA). It is also expensive, as each intubation requires use of a relatively expensive ILMA tracheal tube and each C-trach is designed for only 20 uses.

Evidence on efficacy is confused by publications that use CTrachs both before and after modifications to improve performance, and in common with other optical laryngoscopes is supported only by a relatively small number of uses in genuinely difficult patients.

Optical stylets

Optical stylets are preformed rigid or malleable metal guides containing an optical system (usually fibreoptics) enabling the view from the tip of the device to be transmitted to the proximal end and viewed directly or by attachment of a camera and screen.[109] An appropriate tracheal tube is loaded onto the stylet, which is then introduced into the airway and manipulated to the glottis. The

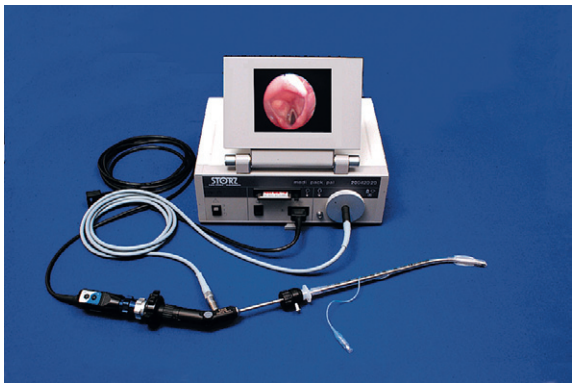


Figure 6.76 The Bonfils laryngoscope with tracheal tube loaded and connected to a compact combination light source and image-processing unit (both by Karl Storz, Germany).

tracheal tube is then advanced over the stylet. While the stylet may be introduced into the trachea this should be avoided, altogether or limited to a small distance to avoid trauma. Oxygen can be insufflated during use, via an attachment for the tracheal tube, in the majority of stylets.

Bonfils intubating fibrescope

The Bonfils intubating fibrescope (Karl Storz, Tuttlingen, Germany) (Fig. 6.76) is a narrow rigid stylet with a gentle (40°) anterior curve at its distal third. It is designed to be introduced from the side of the mouth in a 'retromolar approach' using the non-dominant hand to elevate the jaw and tongue.[110] Other techniques may also be used. An advantage of the Bonfils is that it may be used when mouth opening is very limited and also in awake patients[111] A paediatric version was recently introduced.

There is limited published experience with the Bonfils and what there is is partially conflicting, with studies reporting high and low success rates and variable learning curves.[112]

The Shikani Optical Stylet and Levitan FPS (First Pass Success)

The Shikani Optical Stylet (Clarus Medical, Minneapolis, USA), which has been available for many years, is a semi-malleable optical stylet for use in a similar fashion to the Bonfils intubating fibrescope. It is available in adult (accepting tracheal tubes >5.5 mm ID) and a paediatric size (>3.0 mm ID). The Levitan FPS (First Pass Success) (Clarus Medical, as before) is a similar slightly shorter device (Fig. 6.77). Both devices have somewhat greater natural curves than the Bonfils and both lack rigorous evaluation of performance in easy and difficult airway management.



Figure 6.77 The Levitan FPS (Clarus Medical, USA), mounted on a standard 'ISO green' Timesco laryngoscope handle.
Image courtesy of Timesco of London, UK.

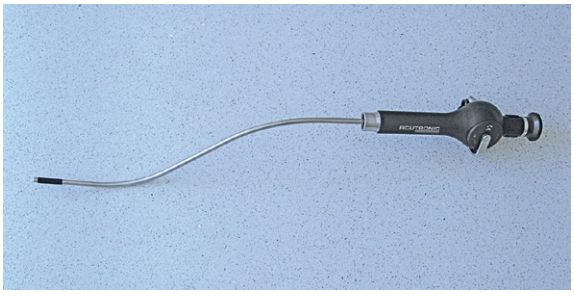


Figure 6.78 SensaScope, Acutronic Medical Systems AG, Switzerland.
Photograph courtesy of LiteOptics, UK.

SensaScope

The SensaScope (Acutronic Medical Systems AG, Switzerland) (Fig. 6.78) is a stylet type rigid fibreoptic laryngoscope with a controllable flexible tip, combining features of both rigid and flexible fibreoptic devices. There is limited, largely positive evaluation of it. A newer video version has just been marketed using a tip mounted LED and CCD thus dispensing with fibreoptics.

Further developments

Developments on optical stylets centre on LED and CCD technology and integrated video screens similar to the McGrath laryngoscope and LMA CTrach, such as the Clarus Video System (Clarus Medical, Minneapolis, USA) [originally: Trachway intubating stylet (Biotronic Instrument Enterprise Ltd, Taiwan, China)].[113] An interesting feature of this device is an additional red LED for transillumination of the larynx (see Trachlight, below). In the light of the arrival of the Ambu aScope (*v.i.*) the fully disposable video stylet will probably precede this publication.

As with many areas of airway management, the profusion of new products militates against careful evaluation and comparisons.

Flexible endoscopes

The first use of flexible fibreoptic technology in airway management can be credited to Dr P. Murphy[114] who, in 1967, reported using the newly invented choledochoscope for intubation of the trachea.[115] Although it was initially ignored by fellow anaesthetists, this approach now forms the mainstay of managing many difficult intubations. The terms flexible intubating fibrescope, endoscope, scope, bronchoscope, laryngoscope, and even tracheoscope are used interchangeably with varying degrees of precision. For anaesthetic purposes it is more important that the device has the correct length, diameter and operating channel for the intended task. Here we use the term flexible fibrescope or scope to describe any suitable device, and flexible endoscope to include also devices not using fibreoptics but video technology for image transmission (see below).

The manoeuvrability of a flexible endoscope means that it can be made to follow virtually any anatomical space and return an image from its tip. Because it presumes no particular anatomical arrangement (i.e. unlike rigid devices there is no pre-shaped curve) and can work around most obstructions, the technique of flexible endoscopic intubation has become a 'gold standard' for management of the difficult laryngoscopy. This has sometimes led to the false assumption that the technique is the solution to all airway difficulties: this is patently not so. However, the versatility and clinical utility of flexible endoscopes make them tools with everyday applications which all anaesthetists should be encouraged to master.

General advantages and limitations/disadvantages include (but are not limited to) the following:

Advantages:

- Primarily, the ability to see round more than one corner due to unparalleled flexibility and manoeuvrability
- Clinical utility: may be used in a very wide variety of clinical circumstances
- Entry of device into glottis under direct vision
- Ability to enter (and examine, anaesthetize, etc.) further into respiratory tree than other airway devices
- Optimal tube position may be confirmed at the time of deployment
- May be used in awake patients
- May be used via oral and nasal routes
- Can be utilized on its own or can make use of any suitable SAD as a conduit to deliver the endoscope to the larynx
- Occasionally the only practical solution to a clinical problem
- In addition to placement of tracheal tubes enables deployment of wires, biopsies, focused sampling of secretions, etc.

Disadvantages:

- Fragile and prone to damage by decontamination procedures and careless or unskilled use
- Requires time to learn and skill to use
- Cost and necessity for decontamination may limit availability and skill acquisition
- Minimum size of tracheal tube used is determined by external diameter of fibrescope (and must be selected at start of procedure)
- Passage of tracheal tube over fibrescope is blind, with risk of failure and trauma
- Difficult to deliver effective oxygenation during use
- Blood and secretions can rapidly render technique ineffective even in skilled hands
- Use in airway obstruction may worsen situation due to 'cork-in-bottle' effect in awake patients
- Difficulty in sterilization may pose risk of transmission of infections (especially working channels).

Flexible fibreoptic laryngoscopes

Principles and design

The pathways through which the illumination and the image pass consist of thousands of very fine glass fibres each typically of the order of 10 μm in diameter. Each fibre consists of a central glass core surrounded by a thin cladding of another type of glass with a different refractive index to that of the core glass. As a result of the difference in refractive indices at the interface of the two materials, light entering the glass fibre undergoes total internal reflection along the length of the fibre to emerge at the other end (Fig. 6.79).

For the purpose of transmitting illumination, the arrangement of the fibres is unimportant, but for image transmission the arrangement of the fibres relative to one another must be identical at either end of the bundle, as each fibre carries a tiny portion of the overall image (in the same way that many small dots make up the printed image in a newspaper): this is called a *coherent bundle*.

The fibres are so fine that they are easily flexible and are lubricated so that they move relative to each other. The whole bundle may therefore be flexed.

The insertion tube of a typical flexible fibreoptic scope (Fig. 6.80) carries:

- two light bundles, running from the 'umbilical cord' to the tip of the scope
- one image bundle, running from objective lens at the tip of the scope to the eyepiece and its lens system
- one 'working' or 'biopsy' channel of a width dictated by the primary purpose of the endoscope
- two opposing angulation wires, which control the more flexible tip of the device bending it in opposite directions within the same plane; angulation is to beyond 90° and often much greater in at least one direction.

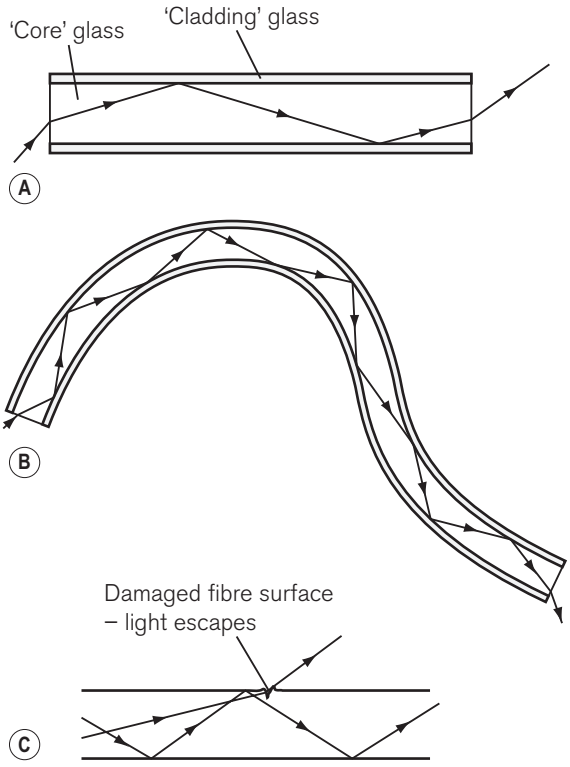


Figure 6.79 **A.** A single optical fibre. Note that the light ray is repeatedly internally reflected from the interface between the core and cladding glass. **B.** If the fibre is curved, the ray is still internally reflected within it. **C.** If the surface of the fibre is damaged, the light ray may not be totally internally reflected and some light may escape from the bundle.

These components are then held together with a stainless steel spiral wrap followed by a stainless steel braid before being covered in a waterproof material to give a rigid cross-section whilst allowing overall flexibility (Fig. 6.81). The fibrescope, whilst able to bend in any plane, is axially rigid to twisting forces, thus rotation of the control handle results in similar rotation of the tip of the device. This forms the basis for control of the tip, which can be rotated so that angulation of the tip, which occurs in only one plane relative to the scope, can be made to take place in any direction.

The fibrescope uses a powerful external cold light source so that the tissues are not damaged by radiant heat. Newer scopes have a detachable light cable which can be replaced by a miniature battery operated light source (Fig. 6.82). These add portability but are not normally powerful enough for use with a video camera ('stack system').

The working channel is used to inject drugs or for suction via the valved port which may be connected to an external vacuum source.

Use

Intubating fibrescopes are available with external diameters ranging from 2.5 mm for use in infants (this smallest size usually has no working channel) through to more than 6 mm, with proportionately larger working channels. Only the channels on the largest sizes (primarily bronchoscopes with working channels over 2 mm) are wide enough for aspirating secretions or blood with any effectiveness; oropharyngeal secretions are best tackled with a handheld Yankauer sucker. A scope with an external diameter of about 3.5 mm is optimal for use in adults (and non-specialist paediatric practice) giving the best combination of:

- stiffness for ease of insertion and
- bore, for access and tracheal tube compatibility.

Video technology

The development of the miniature video camera revolutionized fibreoptic endoscopy (see also under Rigid optical laryngoscopes) by allowing a small camera projecting to a large remote monitor to be attached to the eyepiece of the device. Without doubt flexible fibrescopes should be used with a video monitor rather than with the operator's eye at the eyepiece. Some advantages are reiterated here:

- improved teaching
- greater involvement and understanding from the anaesthetic team leading to improved assistance (e.g. jaw lift is better maintained by someone who can see the effect it has on the laryngeal view enjoyed by the endoscopist)
- the operator has a broader view of the patient's condition if not squinting through an eyepiece and hence a less goal-obsessed approach to endoscopy/intubation
- facility to record images for review and documentation
- operator comfort.

Fig. 6.83 shows the arrangement of components on a typical fibreoptic intubation trolley (Fig. 6.84). Ideally a printer able to capture still images from the outputs of the signal processor or video recorder should also be incorporated with all video systems.

The further development and miniaturization of the CCD (Charge Coupled Device, effectively the 'retina' of modern digital cameras) has allowed a new generation of electronic flexible endoscopes where the CCD is placed at the distal tip of the instrument and the image is transmitted electronically with none of the image loss associated with light transmission along glass fibres (Fig. 6.85). These '*videoscopes*' do not have an eyepiece and are used with a monitor. This CCD arrangement used in conjunction with similarly placed high power LEDs for illumination, obviates the need for fibreoptic bundles completely. Such devices, allied with developments in electronic image processing and screen

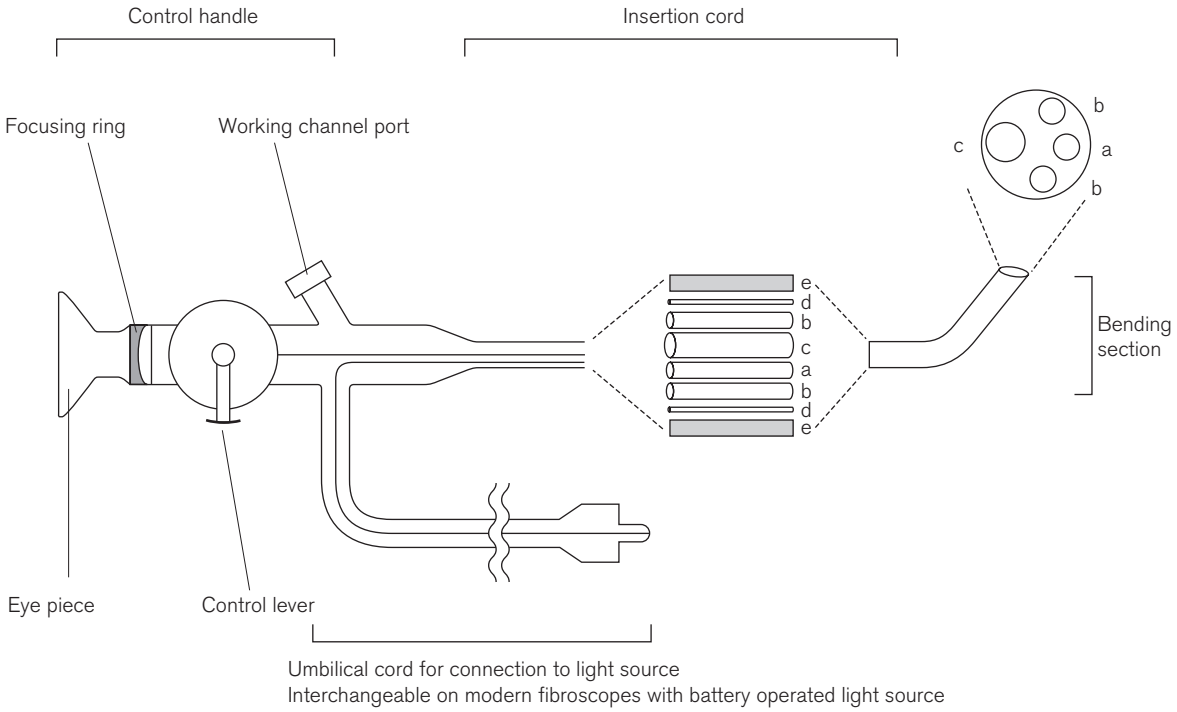


Figure 6.80 Schematic diagram of flexible fibrescope. **a.** Coherent fibreoptic bundle for image transmission; **b.** light transmission bundles for illumination; **c.** working channel; **d.** angulation wires; **e.** outer casing with spiral wrap and flexible steel braid.

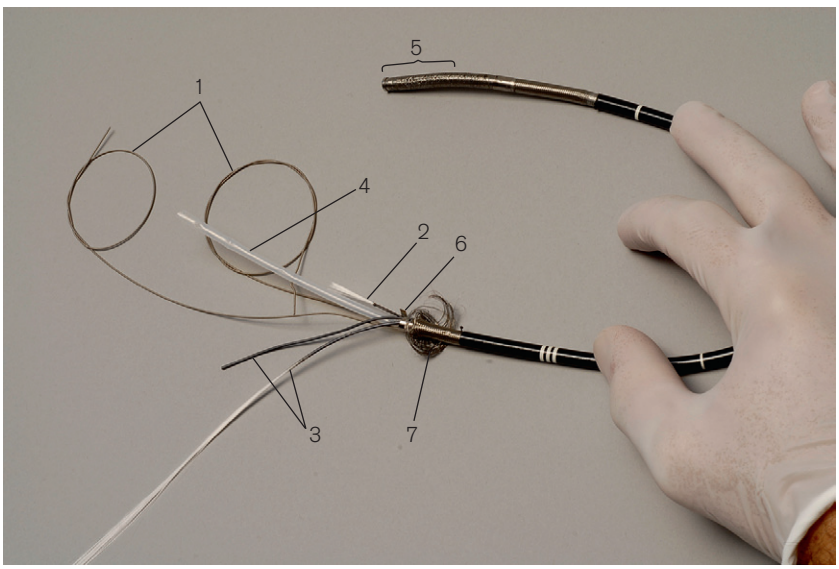


Figure 6.81 Cut away photograph of Olympus fibrescope to show constituent parts. **1,** Angulation wires; **2,** coherent bundle; **3,** light transmission bundles; **4,** working channel; **5,** bending section; **6,** spiral steel wrap; **7,** flexible steel braid.



Figure 6.82 Storz 'portable' flexible intubating fibrescope with choice of battery operated and remote light sources.

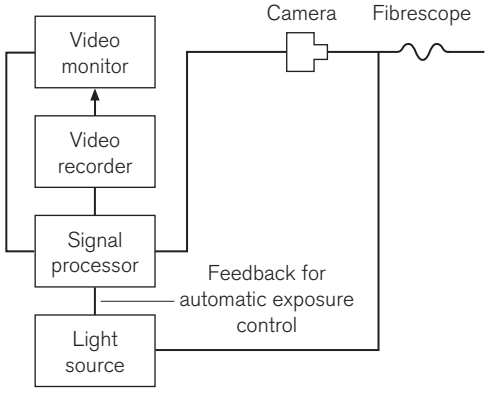


Figure 6.83 Arrangement and interconnection of components for a fibreoptic intubation trolley. The video monitor can toggle between a live image from the camera or playback recorded images.

technology, have dramatically improved the imaging possibilities, currently at a premium in cost.

Ambu aScope

Very recently, a fully disposable, single-patient use, flexible videoscope has been marketed (Ambu aScope, Ambu

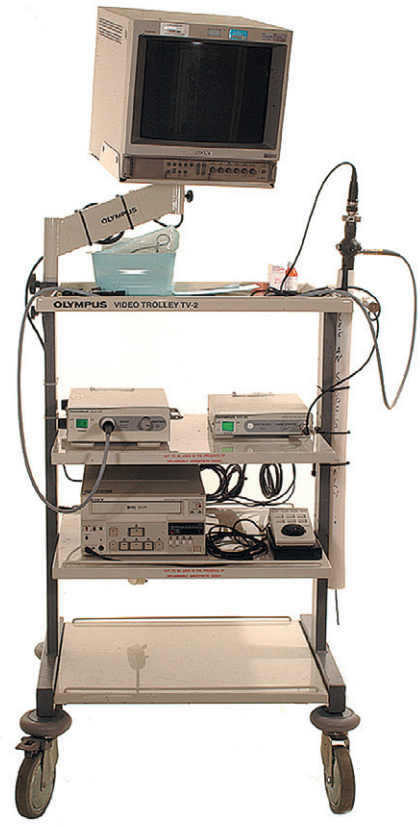


Figure 6.84 A fibreoptic intubation trolley.

A/S, Denmark) which is presented in sterile ready-to-use packaging (Fig. 6.86) with a 3-year shelf life. The device contains no fibreoptics utilizing instead a tip-mounted video camera and dual LEDs (Fig. 6.87), and it is used in conjunction with a battery-operated rechargeable lightweight LCD monitor.

The design concessions to disposability result in a larger diameter insertion cord at 5.2 mm diameter and a diminishing range of tip angulation with increasing flexion along the length of the insertion cord. This can be alleviated, if needed, by holding the part straight and taut. The device is time limited to ensure against reuse: an on/off switch permits half an hour of use within 24 hours of first switch-on. In clinical use, in all other respects the device functions like a traditional flexible fibrescope with a high-quality image. A working channel of 0.8 mm diameter allows topical drug delivery.

When compared to acquisition, decontamination and maintenance costs of traditional fibrescopes, it may be an economical option for infrequent users of endoscopes. There are as yet few trials comparing its performance with traditional fibreoptic or videoscopes.

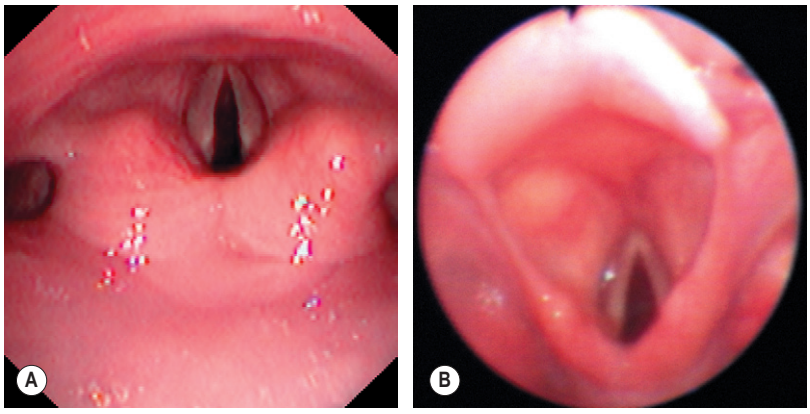


Figure 6.85 Images as seen with **A.** an Olympus digital videoscope; **B.** a fibreoptic system, for comparison of image quality.

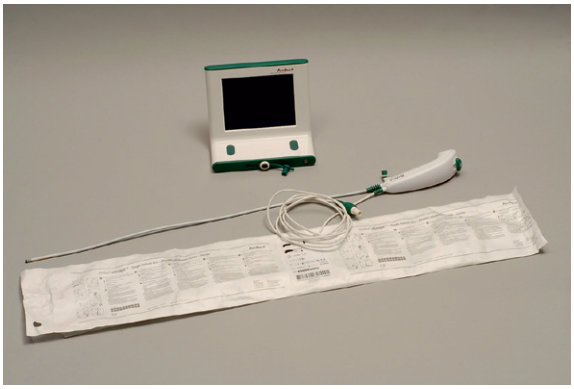


Figure 6.86 The aScope (Ambu A/S, Denmark): a single-patient use intubation videoscope and associated monitor.

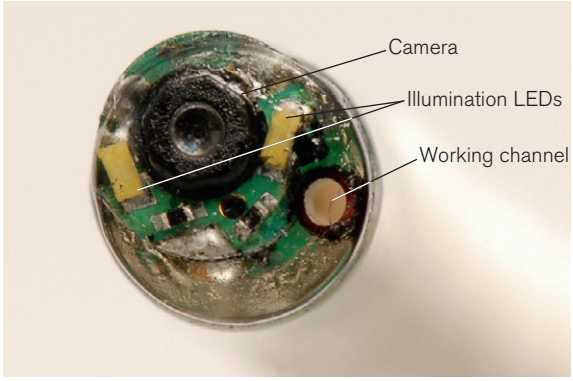


Figure 6.87 The distal (insertion) tip of the Ambu aScope.

Care of the flexible endoscope

(See Endoscope processing – Automated decontamination, Chapter 21). The endoscope must be cleaned and disinfected following use. This has become a lengthy process and is an expensive requirement. Automated reprocessing machines now produce an electronic or, at minimum, a printed paper trail to document completion of the decontamination process.

Modern devices are fully immersible; this has long been denoted on Olympus equipment by a blue ring around the control handle (a red ring denotes an item such as the light cable which may be autoclaved). Some manufacturers even produce fully autoclaveable flexible fibrescopes.

Flexible endoscopic ('fibreoptic') intubation

An in-depth description of techniques and variations is beyond the remit of this chapter.

Technique Flexible endoscopic laryngoscopy via the oral or nasotracheal route may be performed on both the awake or asleep patient. In a patient who is awake, following suitable topical anaesthesia or nerve block, the scope may be inserted, via the nasal or oral route, until it lies behind the tongue and thence advanced into the larynx. The patient may be encouraged to protrude the tongue and/or lower jaw to create the space in the pharynx necessary for the flexible scope to be able to view, identify and advance through the airway. Deep respiration assists the identification of the larynx if needed. Once well in the trachea, a tracheal tube which has been previously loaded onto the endoscope is advanced to lie within the trachea, and the scope is then removed. Once early anxiety is overcome, intubation of the awake rather than anaesthetized patient is often easier as patient co-operation assists the operator and avoids the need for assistants performing manipulations (e.g. jaw and tongue advancement). Sedation may be used for additional patient comfort, but should be light enough to enable patient positioning,

comfort and co-operation. Choice of sedative agent is largely a matter of clinical experience and choice and there is no convincing evidence that use of a low-dose hypnotic (e.g. propofol; anxiolytic, sedative and amnesic) offers better or worse intubation conditions than low-dose opioid infusion (sedation and anti-tussive properties); the combination, however, is likely a poor choice.

Training Competent flexible endoscopic intubation requires training and the maintenance of skills by regular deployment. 'Hand-eye coordination' and the control of the flexible scope may be learnt by most with only brief practice on off-patient models designed to act as 'obstacle courses'. Knowledge of airway anatomy should precede clinical use. Attainment of clinical competence requires about 20 uses[116] and expertise perhaps more than 50.

Associated equipment

Tracheal tubes The ideal tube for intubation over an endoscope would have a narrow kink-resistant wall with a blunt tapered tip that fits the device closely so that it may follow the flexible endoscope easily without snagging on nasal or laryngeal structures.[117] The ILMA tracheal tube has many of the desirable features described and evaluates well in this situation.[118] The Portex Ivory polar tubes (Fig. 6.41) in 6 mm ID size are easy to use and atraumatic in awake patients by virtue of being soft walled with short bevels and are ideally suited for head and neck surgery. Mallinckrodt reinforced ranges with an internal diameter of 6–7 mm, are also popular.

Bite blocks Traditional endoscopy designs are large flanged cylinders placed between the incisors to keep the mouth open. For oral intubation, a dental prop or bite block (Fig. 6.88) ensures that expensive equipment is not inadvertently bitten, is comfortable for the patient and also allows tongue protrusion. A new improved bite block design for anaesthesia (Breathesafe, Fig. 6.5) is an alternative that has a softer more rounded profile. A short handle allows positioning without inserting fingers in the patient's mouth.

Conduit oropharyngeal airways These aim to make oral fibreoptic intubation easier by centring the insertion point in the mouth and delivering the fibrescope closer to the larynx (Fig. 6.4). A number of open or split designs are available which allow the fibrescope to be separated from the airway after tracheal insertion, so as not to limit the size of tracheal tube that may be used for intubation. They also function as a bite block, but they make tongue protrusion, which is a helpful manoeuvre, quite difficult. They are not necessary for awake intubation and for the trainee endoscopist they remove much of the learning that may be gained from an oral intubation. The LMA, as a conduit, is discussed separately below.

Tongue forceps Tongue forceps with atraumatic broad-ridged rubber inserts at the tip (Fig. 6.89) enable tongue retraction to create space in the pharynx for fibreoptic intubation and for injection of local anaesthesia in anaesthetized patients. Manual tongue distraction with a gauze swab is an alternative. The awake or appropriately sedated patient with normal pharyngeal function will protrude the tongue to command, usually obviating the need for this device.



Figure 6.88 Dental props are ideal bite blocks for awake oral intubation as they allow tongue protrusion. The all rubber McKesson pattern on the left are more comfortable.



Figure 6.89 Tongue forceps, surprisingly tolerable.

Ventilation/intubation masks The mask has a large hole within its bowl which is covered by a removable diaphragm. The diaphragm in turn has a small sealable hole through which a fibrescope can be passed (Fig. 6.90). Use of the facemask (usually in an anaesthetized patient) enables oral fibreoptic intubation while delivering oxygen by facemask. Once intubation is complete the expansible properties of the diaphragm allow it to be removed over the tracheal tube. Endoscopy may also be performed through an angle-piece with a silicone rubber diaphragm with an orifice that stretches to accommodate a tracheal tube (Fig. 6.90). Such arrangements may be difficult to use.

Aintree Intubation Catheter (AIC) Developed in Liverpool as 'a ventilation-exchange bougie' and first described in 1996,[119] the AIC enables fibreoptic intubation through the LMA or other SAD. Produced as a single-use device by Cook Medical, it is a 56 cm-long hollow plastic catheter that will fit over flexible endoscopes of 4.5 mm or less diameter, leaving the bending section of the scope free (Fig. 6.91). The assembly is easily passed through the LMA and into the trachea (Fig. 6.92). The scope is then removed, followed by the LMA to leave the AIC in the trachea.

Tracheal tubes of 6.5 mm ID or greater may then be railroaded into place over this. For rescue oxygenation a 15 mm male ISO adapter and female Luer-Lock adapter are both supplied in the AIC package. Great care must be taken if jetting through wide-bore catheters placed close to the carina, for which a much lower driving pressure should be used. The combination of LMA, AIC and flexible endoscope constitutes a powerful technique for airway management because:

1. the standard LMA is a familiar device with a proven track record for establishing (or rescuing) an airway
2. other techniques for intubation through the LMA often fail or require use of impractically long and narrow tracheal tubes, after which the LMA is difficult to remove
3. the cuff of the LMA seals off the larynx from any blood or debris above, thus the conduit protects the flexible scope from its greatest adversaries
4. an absolute minimum of endoscopy skills are needed to rapidly manipulate the flexible scope into the trachea; the technique is thus within reach of almost all anaesthetists.

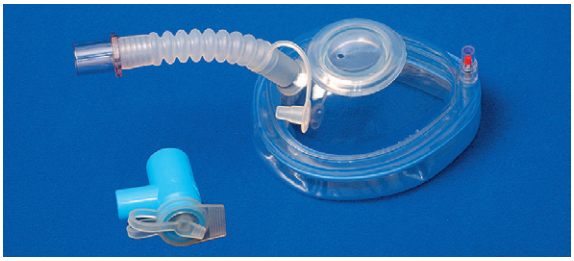


Figure 6.90 Facemask with perforated silicone membrane allowing fibreoptic intubation on the anaesthetized patient, the tube can be forced through the membrane, which is replaceable. The same effect can be achieved using the Mainz Universal Adapter (pictured alongside) and an ordinary facemask.

AIDS FOR INTUBATION/ TUBE EXCHANGE

Bougies and stylets

Not infrequently at direct laryngoscopy, the larynx may be only partially visualized or hidden behind the epiglottis and beyond reach with the normal curvature of a tracheal tube. Intubation may then possibly be accomplished by either:

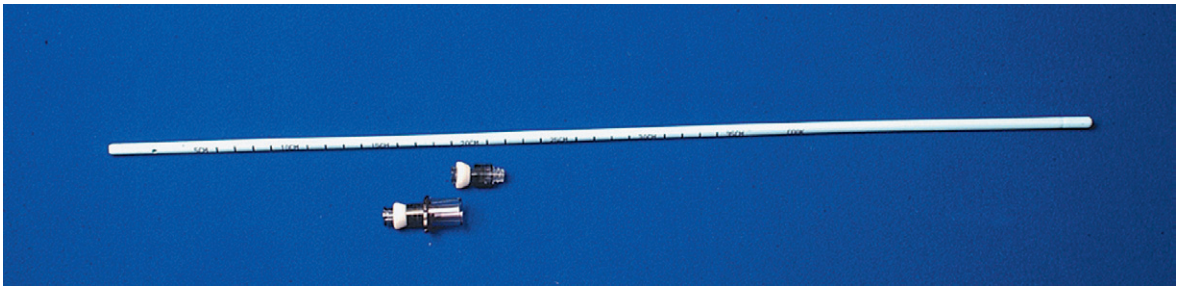


Figure 6.91 The Aintree Intubation Catheter, with Rapi-fit Luer-Lock and ISO 15 mm adaptors.

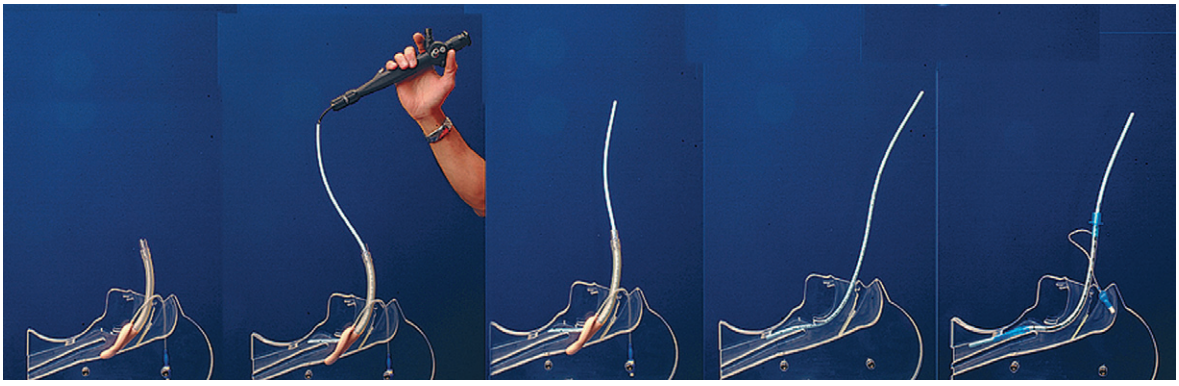
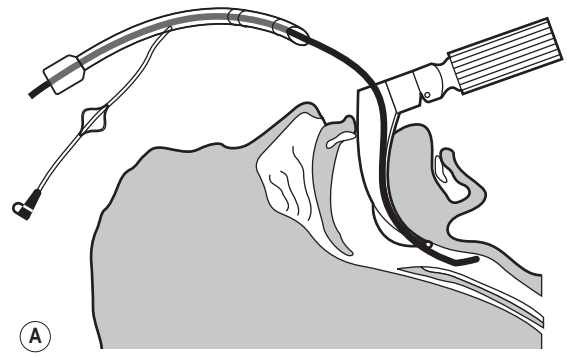


Figure 6.92 Photographic sequence to show fibreoptic intubation through the LMA using the Aintree Intubation Catheter.

- altering the curvature of the tracheal tube using a malleable plastic-coated metal stylet, or
- initially inserting a long thin 'gum-elastic bougie' (GEB) and using this as a guide over which the tube may be passed (rail-roaded) into the trachea (Fig. 6.93).

Credited to Dr P. Venn, the GEB with a coudé tip was designed in the 1970s as the Eschmann Tracheal Tube Introducer (now available as a Portex product, Smiths Medical).[120,121] It has been hugely successful due to the angled tip and the 'memory' of the material for holding a curve (i.e. once curved it will only return to its original shape over a period of 10–20 s, allowing ample time for use). Latto, in his surveys of UK anaesthetic practice, has found the preference for the GEB over other aids at difficult direct laryngoscopy to rise from 45% in 1984 to 100% in 1996 with up to 4% of tracheal intubations being performed with the aid of the GEB.[122]

The gum elastic of the traditional GEB is rendered brittle on repeated autoclaving. The Portex GEB is therefore now marketed as being fit for decontamination and reuse up to five times only.



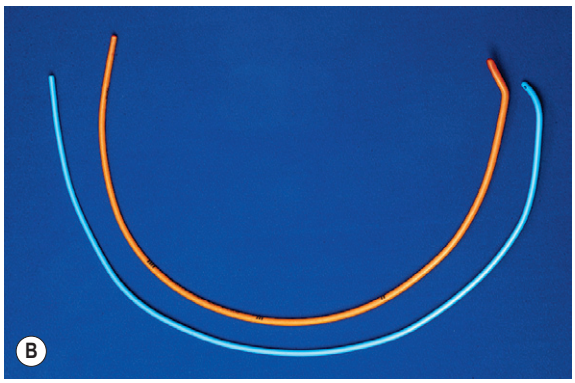


Figure 6.93 **A.** The tracheal tube introducer with coudé tip; **B.** the Eschmann gum-elastic bougie and the blue single-use Frova introducer from Cook.

There are now many new single-use versions of the bougie made from a wide variety of plastics. Some are particularly unsatisfactory. Unfortunately the almost ideal properties of the GEB have proved difficult to replicate: problems include excessive rigidity (risk of trauma), excessive flexibility (poor memory), rapid softening at higher temperatures (floppiness) and the lack of a coudé tip all contributing to poor or dangerous performance characteristics.[123–125]

Several recently introduced bougies offer an additional feature in that being hollow they may be used for oxygen insufflation in the case of difficult or failed intubation. While this is a potentially beneficial additional function, it should not detract from the prime role of an intubation guide: to assist tracheal intubation. In reality many current devices fail in this regard and connectors for attaching oxygen are often poorly prepared. The Frova introducer, despite its undesirable rigidity, has reasonable 'memory' not dissimilar to the GEB (Fig. 6.93B), and can be supplied with a Rapifit adapters with Luer-Lock and ISO 15 mm fittings to allow gas sampling or oxygenation.

Light wand

A logical development from the stylet is the lighted stylet, allowing transillumination of the neck tissues as a further aide to positioning of the tracheal tube during direct laryngoscopy (Fig. 6.94). These devices have limited application now as they have been largely superseded by other developments such as optical stylets that offer both distal illumination and imaging from the device tip. The one area where lighted stylets may have role in skilled hands is when the airway is extensively soiled by blood; in these circumstances the flexible endoscope is often unusable and conventional techniques relying on lines of sight may also fail.

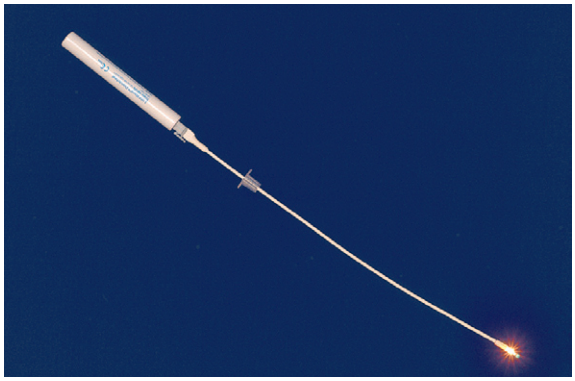


Figure 6.94 The lighted stylet.

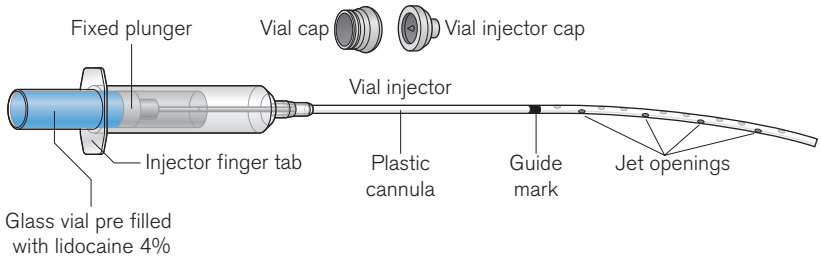


Figure 6.95 The Trachlight prepared for intubation.

Trachlight

The Trachlight (Laerdal Medical) develops the lightwand concept one stage further by aiming to enable intubation using only transillumination of the neck (without laryngoscopy). The stylet is here more rigid than the lightwand to enable some tissue retraction.

A reusable handle houses the battery, to which are attached particularly bright disposable lighted stylets of three different sizes. The tracheal tube is mounted on the stylet assembly and is held by the handle (Fig. 6.95). The stylet is bent into an L-shape before insertion into the mouth. The non-dominant hand grips and elevates the lower incisors and mandible and the Trachlight is rotated into place to transilluminate the thyroid cartilage in the midline. At this point the wire core of the stylet is held steady whilst the rest of the Trachlight and tracheal tube are advanced into the trachea. The device is then disconnected from the tracheal tube and removed.

Intubation in skilled hands appears simple and smooth enough for routine use, with some reporting only a 1% failure rate.[126,127] Sore throat and tracheal bleeding are reported and use in obese patients may not be straightforward.[127,128]

The Trachlight requires skill to distinguish between bright central position illumination (thyroid cartilage) and dull lateral illumination (pyriform fossa). It may require procedures to be performed in a darkened room. The technique is a form of blind intubation.

Notwithstanding these limitations, the light wand and Trachlight may be effective even in the presence of blood in the airway and retain a prominent position in some national airway algorithms.

Airway Exchange Catheter (AEC)

When a tracheal tube must be changed (e.g. to increase its size or change its type), particularly in a patient with an 'at risk' airway, simple extubation and re-intubation may be a poor plan. Tracheal tubes can be more safely exchanged over a catheter or bougie that has been inserted through the tracheal tube into the trachea, the old tube is then removed and the new tube simply railroaded into place over the exchange catheter. The Airway Exchange Catheter (AEC) by Cook Medical (Fig. 6.96), is long enough (83 cm) to enable comfortable removal of an oral or nasal tracheal tube without risk of losing hold of the catheter. Several sizes are available: 11, 14, 19 French gauge, suitable for exchange of tracheal tube 4, 5, 7 mm or greater ID respectively. A paediatric AEC is also available (45 cm long, 8 French, suitable for tubes of 3 mm or greater ID).

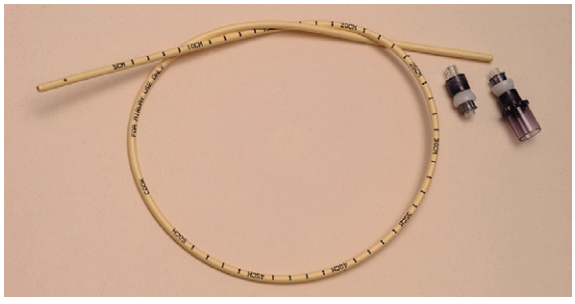


Figure 6.96 The Cook Airway Exchange Catheter with Rapi-Fit adaptors.

Where there may be doubt as to the ability of a postoperative patient (or one sedated on intensive care) to maintain a patent airway following extubation, the AEC may be inserted first and extubation performed 'over' this. It may then be removed some hours later or used to guide re-intubation if needed. Because the plastic remoulds at body temperature and the AEC is smaller than the tracheal tube it replaces it is usually well tolerated by patients.[129] Rapifit connectors are supplied and oxygen insufflation or where necessary very careful jetting is possible as with the Aintree catheter (see above, particularly regarding cautions with this technique). Recently introduced extra firm AECs for use with double lumen tubes use a softer material for the distal 7 cm to reduce the risk of trauma.

Retrograde intubation

Retrograde intubation enables intubation from above, even when laryngoscopy is impossible, by passing a guide up through the larynx from below; this guide is then used to enable railroading of a secondary guide or a tracheal tube into the trachea from above. It is usually performed awake.

A number of techniques are described.[130] In one a cannula is introduced through the cricothyroid membrane with its tip pointing cephalad. A flexible J-tipped guide wire is then passed via the cannula and the tip grasped in the mouth or as it emerges from the nose. A plastic catheter is passed over the J-tip and back along the guide wire to provide a stiffer guide over which the tracheal tube may be railroaded. An epidural needle and catheter are potential emergency alternatives but prepared sets are now made by several manufacturers and if the technique might be contemplated these should ideally be available (Fig. 6.97).

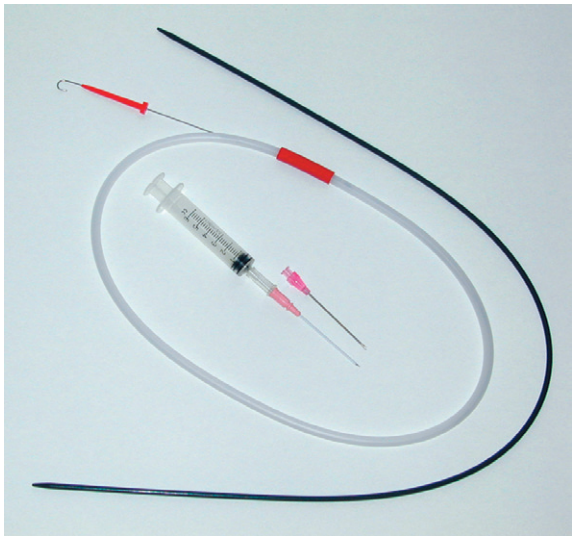


Figure 6.97 A retrograde intubation kit, the principle components of which are a stiff guide wire with a soft flexible J tip and an introducing cannula. A 70 cm 11 Fr Teflon catheter runs over the guide wire to give extra stiffness for railroading the tracheal tube.

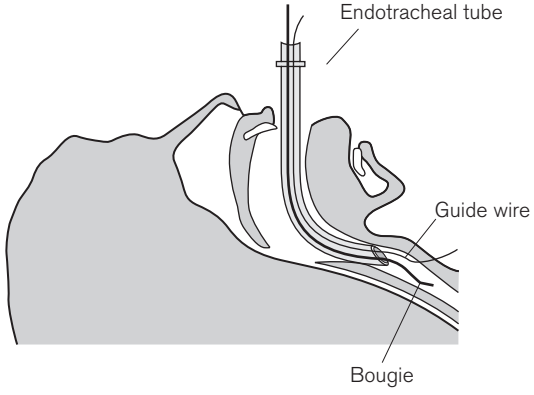


Figure 6.98 To assist in ensuring that the tracheal tube follows down the trachea after removal of the guide wire, a gum elastic bougie can be introduced through the tracheal tube first.

Whichever technique is chosen the very high entry point of the wire into the trachea means that only a short length of tracheal tube can be inserted before it abuts the anterior tracheal wall and the guide wire must be withdrawn (proximally). To ensure being able to insert the tube further into the trachea before withdrawing the wire, a gum elastic bougie may be inserted through the tracheal tube once it is within the trachea (Fig. 6.98), or the emerging guide wire from the larynx may be introduced up the operating channel of a flexible fibrescope with a preloaded tracheal tube and railroaded under direct vision.

MISCELLANY

Magill's forceps

These are ergonomically designed forceps, the handles of which fit comfortably into the operator's right hand like a pair of scissors (Fig. 6.99). The tips are spatulate and ridged for gripping the tip of a tracheal tube so that it can be lifted from the back of the pharynx and into the larynx. They are very useful for assisting nasal intubations, tube exchanges in children and placement of difficult nasogastric tubes.

Drug Delivery Systems

Nebulizing sprays are used for the topical application of local anaesthetic solutions (e.g. 4% lidocaine to the larynx and trachea). The principles of the Forrester spray are shown in Fig. 6.100. When the air reservoir bulb is compressed, the chamber containing the drug solution is pressurised forcing the liquid along a narrow bore delivery tube to the tip of the apparatus. The rest of the air from the bulb is directed to the tip where it mixes with and nebulizes the solution. The sprays tend to block if they are not cleaned shortly after use: the solution remaining at the nozzle dries out, leaving crystals that clog the small orifice. This may be avoided by rinsing out with distilled water or spirit, before cleaning and disinfecting for subsequent usage.

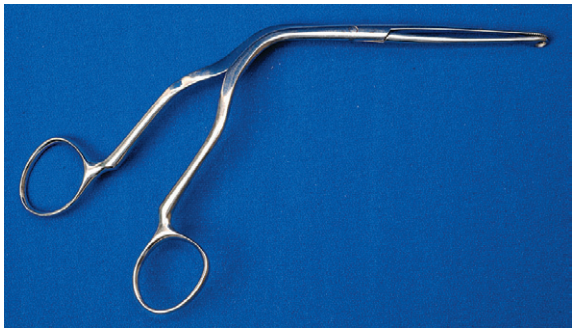


Figure 6.99 Magill's intubating forceps.

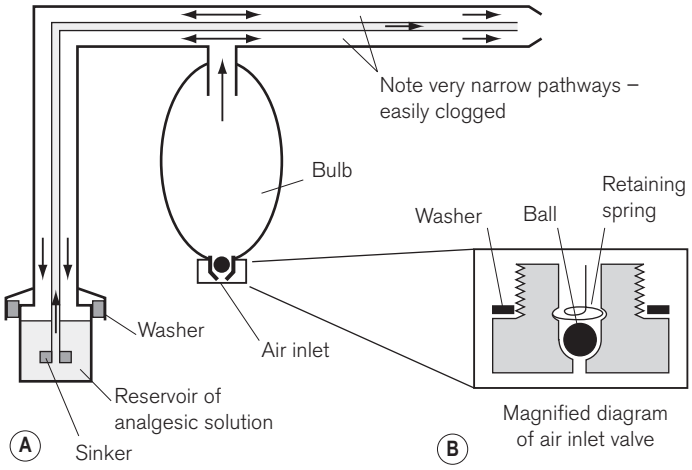


Figure 6.100 The Forrester spray. **A.** Working principles. Note that the diameters of the tubes leading to the nozzles are very small and if analgesic solution is allowed to collect and crystallize out in this area the spray will be blocked. **B.** The air inlet valve.



Figure 6.101 Oxygen powered atomizer.

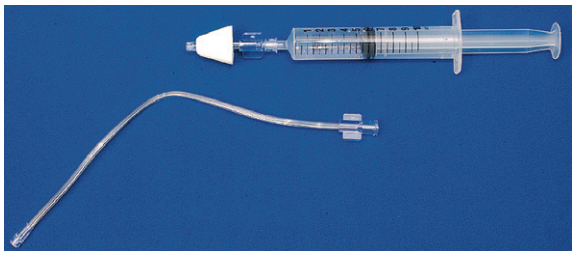


Figure 6.102 The Mucosal Atomization Device, above with syringe attached for intranasal application, and below for intraoral and laryngeal use.

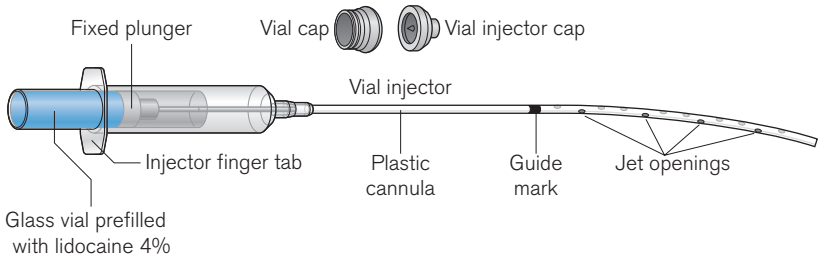


Figure 6.103 Cannula with side holes for delivering drug to the larynx.

The same principles are used in atomizers using pressurized oxygen to drive the nebulization (Fig. 6.101). For use through the flexible scope, single-use dispensers are available with controlled dosing of drug from a small syringe using oxygen as the propellant, they contribute little.

The Mucosal Atomization Device (Wolfe Tory Medical, Salt Lake City, USA) (Fig. 6.102) simply relies on forcing solution through a very narrow orifice for atomization.

The Laryngojet (International Medication Systems Ltd, USA) is a prefilled drug delivery system for 4% lidocaine topicalization of the airway. The packaging system will be familiar from the emergency drugs often presented in resuscitation kits. A prefilled glass cylinder has a perforated rubber stopper which when inserted into the vial injector becomes the fixed plunger of the syringe. Here the injector has a long plastic cannula with side holes for delivering drug to the larynx (Fig 6.103).

REFERENCES

1. Verghese C. Airway management. *Curr Opin Anaesthesia* 1999;**12**: 667–74.
2. http://www.mhra.gov.uk/Safetyinformation/Generalsafetyinformationandadvice/Technicalinformation/PhthalatesinPVCmedicaldevices/index.htm. Accessed 30/05/10
3. Thomas DI. Overcoming the beard. *Anaesthesia* 1999;**54**:100.
4. Vincent C, Ames WA. The bearded airway. *Anaesthesia* 1998;**53**: 1034–35.
5. Brain AI. The laryngeal mask – a new concept in airway management. *Br J Anaesth* 1983; **55**:801–5.
6. Frass M, Frenzer R, Mayer G, Popovic R, Leithner C. Mechanical ventilation with the esophageal tracheal combitube (ETC) in the intensive care unit. *Arch Emerg Med* 1987;**4**:219–25.
7. Jarquin-Valdivia AA, Thompson A. The origin of the combitube. *Int J Anesth* 2002;5.
8. Ocker H, Wenzel V, Schmucker P, Steinfath M, Dörges V. A comparison of the laryngeal tube with the laryngeal mask airway during routine surgical procedures. *Anaesth Analg* 2002; **95**:1094–97.
9. Asai T, Kawashima A, Hidaka I, Kawachi S. The laryngeal tube compared with the larygeal mask: insertion, gas leak pressure and gastric insufflation. *Br J Anaesth* 2002;**89**:729–32.
10. Brimacombe J. A proposed classification system for extraglottic airway devices. *Anesthesiology* 2004; **101**:559
11. Miller DM. A proposed classification and scoring system for supraglottic sealing airways: a brief review. *Anesth Analg* 2004; **99**:1553–9.
12. Keller C, Brimacombe J, Rädler C, Pühringer F. Do laryngeal mask airway devices attenuate liquid flow between the esophagus and pharynx? A randomized, controlled cadaver study. *Anesth Analg* 1999;**88**:904–7.
13. Keller C, Brimacombe J, Kleinsasser A, Loeckinger A. Does the ProSeal laryngeal mask airway prevent aspiration of regurgitated fluid? *Anesth Analg* 2000;**91**:1017–20.
14. Schmidbauer W, Bercker S, Volk T, Bogusch G, Mager G, Kerner T. Oesophageal seal of the novel supralaryngeal airway device I-Gel in comparison with the laryngeal mask airways Classic and ProSeal using a cadaver model. *Br J Anaesth* 2009;**102**:135–9.
15. Bercker S, Schmidbauer W, Volk T, Bogusch G, Bubser HP, Hensel M, Kerner T. A comparison of seal in seven supraglottic airway devices using a cadaver model of elevated esophageal pressure. *Anesth Analg* 2008;**106**:445–8.
16. Brimacombe J. Analysis of 1500 laryngeal mask uses by one anaesthetist in adults undergoing routine anaesthesia. *Anaesthesia* 1996;**51**:76–80.
17. Abdi W, Dhonneur G, Amathieu R, Adhoum A, Kamoun W, Slavov V, et al. LMA Supreme(tm) Versus Facemask Ventilation Performed by Novices: A Comparative Study in Morbidly Obese Patients Showing Difficult Ventilation Predictors. *Obes Surg* 2009;**19**(12):1624–30.
18. Kapila A, Addy EV, Verghese C, Brain AI. The intubating laryngeal mask airway: an initial assessment of performance. *Br J Anaesth* 1997; **79**:710–13.
19. Brain AI, Verghese C, Addy EV, Kapila A, Brimacombe J. The intubating laryngeal mask. II: A preliminary clinical report of a new means of intubating the trachea. *Br J Anaesth* 1997;**79**: 704–9.
20. Caponas G. Intubating laryngeal mask airway. *Anaesth Intensive Care* 2002;**30**:551–69.
21. Pandit JJ, MacLachlan K, Dravid RM, Popat MT. Comparison of times to achieve tracheal intubation with three techniques using the laryngeal or intubating laryngeal mask airway. *Anaesthesia* 2002;**57**:128–32.
22. McNeillis NJ, Timberlake C, Avidan MS, Sarang K, Choyce A, Radcliffe JJ. Fibreoptic views through the laryngeal mask and the intubating laryngeal mask. *Eur J Anaesth* 2001;**18**: 471–5.
23. Keller C, Brimacombe J, Puhringer F. A fibreoptic scoring system to assess the position of laryngeal mask airway devices. Interobserver variability and a comparison between the standard, flexible and intubating laryngeal mask airways. *Anaesthesia* 2000;**35**: 692–4.
24. Kihara S, Yaguchi Y, Brimacombe J, Watanabe S, Taguchi N. Routine use of the intubating laryngeal mask airway results in increased upper airway morbidity. *Can J Anaesth* 2001;**48**: 604–8.
25. Asai T, Shingu K. The laryngeal tube. *Br J Anaesth* 2005;**95**: 729–36.
26. Cook TM, BMcCormick, Asai T. Randomized comparison of laryngeal tube with classic laryngeal mask airway for anaesthesia with controlled ventilation. *Br J Anaesth* 2003; **91**:373–8.
27. Hooshangi H, Wong DT. Brief review: the Cobra Perilaryngeal Airway (CobraPLA) and the Streamlined Liner of Pharyngeal Airway (SLIPA) supraglottic airways. *Can J Anaesth* 2008; **55**:177–85.
28. van Zundert A, Al-Shaikh B, Brimacombe J, Koster J, Koning D, Mortier EP. Comparison of three disposable extraglottic airway devices in spontaneously breathing adults: the LMA-Unique, the Soft Seal laryngeal mask, and the Cobra perilaryngeal airway. *Anesthesiology* 2006;**104**:1165–9.
29. Cook TM, Lowe JM. An evaluation of the CobraPLA. Stopped after two cases of aspiration. *Anaesthesia* 2005;**60**:791–6.
30. Farrow C, Cook TM. Pulmonary aspiration through a Cobra PLA. The Cobra bites back. *Anaesthesia* 2004;**59**:1140–1.

31. Brain AI, Verghese C, Strube PJ. The LMA 'ProSeal' – a laryngeal mask with an oesophageal vent. *Br J Anaesth* 2000;**84**:650–4.
32. Cook TM, Lee G, Nolan JP. The ProSealTM laryngeal mask airway: a review of the literature. *Can J Anaesth* 2005;**52**:739–60.
33. Kelly FK, Sale S, Bayley G, Cook TM, Stoddart P, White MC. A cohort evaluation of the Paediatric Proseal laryngeal mask airway in 100 unpremedicated children. *Pediatr Anaesthesia* 2008;**19**:947–51.
34. Howath A, Brimacombe J, Keller C. Gum-elastic bougie-guided insertion of the ProSeal laryngeal mask airway: a new technique. *Anaesth Intensive Care* 2002;**30**: 624–7.
35. Brimacombe J, Keller C, Judd DV. Gum elastic bougie-guided insertion of the ProSeal laryngeal mask airway is superior to the digital and introducer tool techniques. *Anesthesiology* 2004; **100**:25–9.
36. Cook TM, Gibbison B. Analysis of 1000 consecutive uses of the ProSeal laryngeal mask airway by one anaesthetist at a district general hospital. *Br J Anaesth* 2007; **99**:436–9.
37. Cook TM, Silsby J, Simpson TP. Airway rescue in acute upper airway obstruction using a ProSeal laryngeal mask airway and an Aintree catheter – a review of the ProSeal laryngeal mask airway in the management of the difficult airway. *Anaesthesia* 2005;**60**: 1129–36.
38. Cook TM, Gatward JJ, Handel J, Hardy R, Thomson C, Shrivastava R, Clarke PA. Evaluation of the LMA Supreme in 100 non-paralysed patients. *Anaesthesia* 2009;**64**:555–62.
39. Theiler LG, Kleine-Brueggeney M, Kaiser D, Urwyler N, Luyet C, Vogt A, Greif R. Crossover comparison of the laryngeal mask supreme and the i-gel in simulated difficult airway scenario in anesthetized patients. *Anesthesiology* 2009;**111**: 55–62.
40. Gatward JJ, Cook TM, Seller C, Simpson T, Vanek V, Handel J, Kelly F. Evaluation of the i-gel airway in 100 patients. *Anaesthesia* 2008;**63**:1124–30.
41. Wharton NM, Gibbison B, Gabbott D, Haslam GM, Cook TM. i-gel insertion by novices in manikins and patients. *Anaesthesia* 2008;**63**:991–5.
42. Bercker S, Schmidbauer W, Volk T, Bogusch G, Bubser HP, Hensel M, Kerner T. A comparison of seal in seven supraglottic airway devices using a cadaver model of elevated esophageal pressure. *Anesth Analg* 2008;**106**:445–8.
43. Gibbison B, Cook TM, Seller C. Case series: protection from aspiration and failure of protection from aspiration with the i-gel airway. *Br J Anaesth* 2008;**100**: 415–17.
44. Vézina D, Lessard MR, Bussières J, Topping C, Trépanier CA. Complications assosciated with the Esophageal-Tracheal Combitube. *Can J Anaesth* 1998;**45**:76–80.
45. Klein, H, Williamson M, Sue-Ling HM, Vucevic M, Quinn AC. Esophageal rupture associated with the use of the Combitube. *Anaesth Analg* 1997;**85**:937–9.
46. Cook TM, Hommers C. New airways for resuscitation? *Resuscitation* 2006;**69**:371–87.
47. Cook TM, Cranshaw J. Randomised controlled crossover trial of the laryngeal mask airway Proseal with the laryngeal tube sonda during anaesthesia with controlled ventilation. *Br J Anaesth* 2005;**95**: 261–6.
48. Mihai R, Knottenbelt G, Cook TM. Evaluation of the LTS II in 100 patients. *Br J Anaesth* 2007;**99**: 734–9.
49. Miller DM, Light D. Laboratory and clinical comparisons of the Streamlined Liner of the Pharynx Airway(SLIPAtm) with the laryngeal mask airway. *Anaesthesia* 2003;**58**:136–42.
50. Miller DM, Camporota L. Advantages of ProSeal and SLIPA airways over tracheal tubes for gynecological laparoscopies. *Can J Anesth* 2006;**53**:188–93.
51. Hein C, Owen H, Plummer J. Randomized comparison of the SLIPA (Streamlined Liner of the Pharynx Airway) and the SS-LM (Soft Seal Laryngeal Mask) by medical students. *Emerg Med Australas* 2006;**18**:478.
52. Lange M, Smul T, Zimmermann P, Kohlenberger R, Roewer N, Kehl F. The effectiveness and patient comfort of the novel streamlined pharynx airway liner (SLIPA) compared with the conventional laryngeal mask airway in ophthalmic surgery. *Anesth Analg* 2007;**104**:431–4.
53. Vamadevan S, Diba A. The Conventional LMA(r): benchmark for assessing supralaryngeal airway efficacy? *Anesth Analg* 2007;**105**: 1508.
54. Gillespie NA. Endotracheal anaesthesia. USA: University of Wisconsin Press; 1941. p. 6–25.
55. Vesalius A. De Humanis Corporis Fabrica. 1st ed. 1542. p. 658.
56. Trendelenburg F. Tamponnade der Trachea. *Arch f Klin Chir* 1871; **12**:121.
57. Waters RM, Rovenstine EA,Guedel AE. Endotracheal anaesthesia and its historical development. *Anaesth Analg* 1933;**12**:196.
58. MacEwen W. Clinical observations on the introduction of tracheal tubes by the mouth, instead of performing tracheotomy or laryngotomy. *Br Med J* 1880;**2**: 122,163.
59. Meltzer SJ, Auer J. Continuous respiration without respiratory movements. *J Exp Med* 1909;**11**: 622.
60. Koh KF, Hare JD, Calder I. Small tubes revisited. *Anaesthesia* 1998; **53**(1):46–50.
61. Stenqvist O, Sonander H, Nilson K. Small tracheal tubes. Ventilator and intratracheal pressures during controlled ventilation. *Br J Anaesth* 1979;**51**(4):375–80.
62. Stout DM, Bishop MJ, Dwersteg JF, Cullen FC. Correlation of tracheal tube size with sore throat and hoarseness following general anaesthesia. *Anaesthesiol* 1987; **67**:419–21.
63. Avrahami E, Frishman E, Spierer I, Englender M, Katz R. CT of minor intubation trauma with clinical correlation. *Eur J Radiol* 1995; **20**:68–71.
64. Shapiro M, Wilson RK, Casar G, Bloom K, Teague RB. Work of

breathing through different sized tracheal tubes. *Crit Care Med* 1986; **14**:1028–31.
65. Nunn JF, Ezi-Ashi TI. The respiratory effects of resistance to breathing in anaesthetised man. *Anaesthesiol* 1961;**22**:174–85.
66. Thornton HL. Vinyl-'Portex' tubing. *Br Med J* 1944;**Jul 1**:14.
67. Tu HN, Saidi N, Leiutaud T, Bensaid S, Menival V, Duvaldestin P. Nitrous oxide increases endotracheal cuff pressure and the incidence of tracheal lesions in anesthetized patients. *Anesth Analg* 1999;**89**:187–90.
68. al-Shaikh B, Jones M, Baldwin F. Evaluation of pressure changes in a new design tracheal tube cuff, the Portex Soft Seal, during nitrous oxide anaesthesia. *Br J Anaesth* 1999;**83**:805–6.
69. Reducing the risk of retained throat packs after surgery. NPSA 28 April 2009 http://www.nrls.npsa.nhs.uk/resources/?entryid45=59853
70. Cook TM, Alexander R. Major complications during anaesthesia for elective laryngeal surgery in the United Kingdom: a national survey of the use of high pressure-source ventilation. *Br J Anaesth* 2008; **101**:266–72.
71. Jaquet Y, Monnier P, Van Melle G, Ravussin P, Spahn DR, Chollet-Rivier M. Complications of different ventilation strategies in endoscopic laryngeal surgery: a 10-year review. *Anesthesiology* 2006;**104**:52–9.
72. Bourgain JL, Desruenne E, Fischler M, Ravussin P. Safety and jet ventilation. *Br J Anaesth* 2008; **101**:573; author reply 574.
73. Andrzejowski J, Francis G. The efficacy of lidocaine administered via the LITA tracheal tube in attenuating the extubation response in beta-blocked patients following craniotomy. *Anaesthesia* 2002;**57**:399–401.
74. Diachun CA, Tuninck BP, Brock-Utne JG. Suppression of cough during emergence from general anaesthesia: laryngotracheal lidocaine through a modified tracheal tube. *J Clin Anaesth* 2001;**13**:447–51.
75. Berra L, De Marchi L, Yu ZX, Laquerriere P, Baccarelli A, Kolobow T. Tracheal tubes coated with antiseptics decrease bacterial colonization of the ventilator circuits, lungs, and tracheal tube. *Anesthesiology* 2004;**100**:1446–56.
76. Shorr AF, Zilberberg MD, Kollef M. Cost-effectiveness analysis of a silver-coated tracheal tube to reduce the incidence of ventilator-associated pneumonia. *Anesthesiology* 2009;**30**:759–63.
77. Ciaglia P, Firsching R, Syniec C. Elective percutaneous dilatational tracheostomy: a new simple bedside procedure – preliminary report. *Chest* 1985;**87**:715–19.
78. Susanto I. Comparing Percutaneous tracheostomy with open surgical tracheostomy: both will coexist until robust evidence becomes available [editorial]. *Br Med J* 2002;**324**(7328):3–4.
79. Delaney A, Bagshaw S, Nalos M. Percutaneous dilatational tracheostomy versus surgical tracheostomy in critically ill patients: a systematic review and meta-analysis. *Crit Care* 2006; **10**:R55.
80. Kluge S, Baumann HJ, Maier C, Klose H, Meyer A, Nierhaus A, Kreymann G. Tracheostomy in the intensive care unit: a nationwide survey. *Anesth Analg* 2008;**107**: 1639.
81. Griggs WM, Worthley LIG, Gilligan JE, Thomas PD, Myburg JA. A simple percutaneous tracheostomy technique. *Surg Gynecol Obstet* 1990;**170**:543–5.
82. Fikkers BG, van Heerbeek N, Krabbe PF, Marres HA, van den Hoogen FJ. Percutaneous tracheostomy with the guide wire dilating forceps technique: presentation of 171 consecutive patients. *Head Neck* 2002;**24**: 625–31.
83. Frova G, Quintel M. A new simple method for percutaneous tracheostomy: controlled rotating dilating. A preliminary report. *Intensive Care Med* 2002;**28**: 299–303.
84. Zgoda MA, Berger R. Balloon-facilitated percutaneous dilational tracheostomy tube placement: preliminary report of a novel technique. *Chest* 2005;**128**: 3688–90.
85. Fantoni A, Ripamonti D. A non-derivative, non-surgical tracheostomy: the translaryngeal method. *Intensive Care Med* 1997; **23**:386–92.
86. Diba A. Transtracheal jet ventilation. *Anaesth Intensive Care Med* 2002;**3**:207–9.
87. Dollner R, Verch M, Schweiger P, Graf B, Wallner F. Long-term outcome after Griggs tracheostomy. *J Otolaryngol* 2002;**31**:386–9.
88. Layman PR. Transtracheal ventilation in oral surgery. *Annu Rev Coll Surg Engl* 1983; **65**:318–20.
89. Benumof JL, Scheller MS. The importance of transtracheal jet ventilation in the management of the difficult airway. *Anaesthesiol* 1989;**71**:769–78.
90. Patel C, Diba A. Measuring tracheal airway pressures during transtracheal jet ventilation: an observational study. *Anaesthesia* 2004;**59**:248–51.
91. Ghani AC, Donnelly CA, Ferguson NM, Anderson RM. Updated projections of future vCJD deaths in the UK. *BMC Infect Dis* 2003; **3**:4.
92. Blunt MC, Burchett KR. Variant Creutzfeldt–Jakob disease and disposable anaesthetic equipment – balancing the risk. *BJA* 2003; **90**:1–3.
93. Stevenson M, Oakley J, Chick S. Patient safety and reduction of risk of transmission of variant Creutzfeldt–Jakob disease via interventional procedures. *NICE* June 2006.
94. Twigg SJ, McCormick B, Cook TM. Randomized evaluation of the performance of single-use laryngoscopes in simulated easy and difficult intubation. *Br J Anaesth* 2003;**90**:8–13.
95. Evans A, Vaughan RS, Hall JE, Mecklenburgh J, Wilkes AR. A comparison of the forces exerted during laryngoscopy using disposable and non-disposable laryngoscope blades. *Anaesthesia* 2003;**58**:869–73.
96. Centre for Evidence Based Purchasing. Buyer's guide: Laryngoscopes. CEP08048, NHS Purchasing and Supply Agency. July 2009.

97. Levitan RM, Ochroch EA. Explaining the variable effect on laryngeal view obtained with the McCoy laryngoscope. *Anaesthesia* 1999;**54**:599–601.
98. Uchida T, Hikawa Y, Saito Y, Yasuda K. The McCoy levering laryngoscope in patients with limited neck extension. *Can J Anesth* 1997;**44**:674–6.
99. Chisholm DG, Calder I. Experience with the McCoy laryngoscope in difficult laryngoscopy. *Anaesthesia* 1997;**52**:906–8.
100. Leon O, Benhamou D. Improvement of glottis visualization with a McCoy blade. *Ann Fr Anesth Reanim* 1998;**17**: 68–71.
101. Cook TM, Tuckey JP. A comparison between the Macintosh and the McCoy laryngoscope blades. *Anaesthesia* 1996;**51**: 977–80.
102. Satava, RM, Gooden SM. The impact of video endoscopy on surgical training. *Am Surg* 1989; **55**:263–66.
103. Mihai R, Blair E, Kay H, Cook TM. A quantitative review and meta-analysis of performance of non standard laryngoscopes and rigid fibreoptic intubation aids. *Anaesthesia* 2008;**63**: 745–60.
104. Pearce AC, Shaw S, Macklin S. Evaluation of the Upsherscope. A new rigid fibrescope. *Anaesthesia* 1996;**51**(6):561–4.
105. Fridrich P, Frass M, Krenn CG, et al. The UpsherScope in routine and difficult airway management: a randomized, controlled clinical trial. *Anesth Analg* 1997;**85**(6): 1377–81.
106. Foley LJ, Ochroch EA. Bridges to establish an emergency airway and alternate intubating techniques. *Crit Care Clin* 2000;**16**:429–44.
107. Low D, Healy D, Rasburn N. The use of the BERCI DCI Video Laryngoscope for teaching novices direct laryngoscopy and tracheal intubation. *Anaesthesia* 2008;**63**: 195–201.
108. Kaplan MB, Hagberg CA, Ward DS, Brambrink A, Chhibber AK, Heidegger T, et al. Comparison of direct and video-assisted views of the larynx during routine intubation. *J Clin Anesth* 2006;**18**:357–62.
109. Liem EB, Bjoraker DG, Gravenstein D. New options for airway management: intubating fibreoptic stylets. *Br J Anaesth* 2003;**91**: 408–18.
110. Wong P, Lawrence C, Pearce A. Intubation times for using the Bonfils intubation fibrescope. *Br J Anaesth* 2003;**91**:757.
111. Corbanese U, Possamai C. Awake intubation with the Bonfils fibrescope in patients with difficult airway. *Eur J Anaesthesiol* 2009;**26**: 837–4.
112. Halligan M, Charters P. A clinical evaluation of the Bonfils intubation fibrescope. *Anaesthesia* 2003;**58**:1087–91.
113. Ong J, Lee CL, Lai HY, Lee Y, Chen TY, Shyr MH. A new video intubating device: Trachway intubating stylet. *Anaesthesia* 2009; **64**:1145.
114. Calder I, Pearce A, Towey R. Classic paper: a fibreoptic endoscope used for tracheal intubation. *Anaesthesia* 1996;**51**:602.
115. Murphy P. A fibreoptic endoscope used for tracheal intubation. *Anaesthesia* 1967;**22**:489–91.
116. Smith JE, Jackson AP, Hurdley J, Clifton PJ. Learning curves for fibreoptic nasotracheal intubation when using the endoscopic video camera. *Anaesthesia* 1997;**52**: 101–6.
117. Jones HE, Pearce AC, Moore P. Fibreoptic intubation. Influence of tracheal tube tip design. *Anaesthesia* 1993;**48**:672–4.
118. Bhatnagar S, Mishra S, Jha RR, Singhal AK, Bhatnagar N. The LMA Fastrach facilitates fibreoptic intubation in oral cancer patients. *Can J Anaesth* 2005;**52**:641–5.
119. Atherton DPL, O'Sullivan E, Lowe D, Charters P. A ventilation-exchange bougie for fibreoptic intubations with the laryngeal mask airway. *Anaesthesia* 1996; **51**:1123–6.
120. Venn PH. The gum elastic bougie. *Anaesthesia* 1993;**48**:274–5.
121. Henderson JJ. Development of the 'gum-elastic bougie'. *Anaesthesia* 2003;**58**:103–4.
122. Latto IP, Stacey M, Mecklenburgh J, Vaughan RS. Survey of the use of the gum elastic bougie in clinical practice. *Anaesthesia* 2002;**57**: 379–84.
123. Mingo O, Suaris P, Charman S, Ferguson C, Patel A. The effect of temperature on bougies: a photographic and manikin study. *Anaesthesia* 2008;**63**:1135–8.
124. Annamaneni R, Hodzovic I, Wilkes AR, Latto IP. A comparison of simulated difficult intubation with multiple-use and single-use bougies in a manikin. *Anaesthesia* 2003;**58**:45–9.
125. Marfin AG, Pandit JJ, Hames KC, Popat MT, Yentis SM. Use of the bougie in simulated difficult intubation. 2. Comparison of single-use bougie with multiple-use bougie. *Anaesthesia* 2003; **58**:852–5.
126. Hung OR, Pytka MD, Morris I, Murphy M, Launcelott G, Stevens S, et al. Clinical trial of a new lightwand device (Trachlight) to intubate the trachea. *Anesthesiology* 1995;**83**:509–14.
127. Tsutsui T, Setoyama K. A clinical evaluation of blind orotracheal intubation using Trachlight in 511 patients. *Anesthesiology* 2001; **50**:854–8.
128. Nishiyama T, Matsukawa T, Hanaoka K. Optimal length and angle of a new lightwand device (Trachlight). *J Clin Anaesth* 1999; **11**:332–5.
129. Loudermilk EP, Hartmannsgruber M, Stoltzfus DP, Langevin PB. A prospective study of the safety of tracheal extubation using a pediatric airway exchange catheter for patients with a known difficult airway. *Chest* 1997;**111**:1660–5.
130. Dhara SS. Retrograde tracheal intubation. *Anaesthesia* 2009;**64**:1094–104.

FURTHER READING

Bercker S, Schmidbauer W, Volk T, Bogusch G, Bubser HP, Hensel M, et al. A comparison of seal in seven supraglottic airway devices using a cadaver model of elevated esophageal pressure. *Anesth Analg* 2008;**106**: 445–8.

Buyers guide. Laryngeal masks. Centre for Evidence Based Purchasing. *NHS Purchasing and suppliers agency. July 2008*.

Buyers guide. Laryngoscopes. Centre for Evidence Based Purchasing. *NHS Purchasing and suppliers agency. CEP08048 July 2009*.

Calder I, Pearce A, editors. Core topics in airway management. Cambridge: University Press; 2005.

Cook TM. Novel airway devices: spoilt for choice? [editorial] *Anaesthesia* 2003;**58**:107–10.

Cook TM, Lee G, Nolan JP. The ProSealT™ laryngeal mask airway: a review of the literature. *Can J Anaesth* 2005;**52**:739–60.

Diba A. Transtracheal jet ventilation. *Anaesth Intensive Care Med* 2002; **3**:207–9.

Jaeger JM, Durbin CG. Special purpose tracheal tubes. *Respir Care* 1999; **44**:661–83.

Latto IP, Vaughn RS, editors. Difficulties in tracheal intubation. 2nd ed. London: W.B. Saunders; 1997.

Popat M. Practical fibreoptic intubation. Oxford: Butterworth-Heinemann; 2001.

Schmidbauer W, Bercker S, Volk T, Bogusch G, Mager G, Kerner T. Oesophageal seal of the novel supralaryngeal airway device i-gel in comparison with the laryngeal mask airways Classic and ProSeal using a cadaver model. *Br J Anaesth* 2009;**102**:135–9.

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Chapter 7

Equipment for the inhalation of oxygen and other gasses

Thomas DA Standley and Daniel W Wheeler

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INTRODUCTION

The administration of supplemental oxygen is a fundamental part of the treatment of the acutely ill patient and those undergoing surgery. The equipment required has evolved to allow its use in a wide variety of circumstances.

The benefits of supplemental oxygen are:

- as a treatment for hypoxaemia due to hypoventilation, or due to gas transfer or ventilation/perfusion abnormalities (e.g. heart failure, anaesthesia)
- to improve oxygen supply to tissues, when a disease process causes oxygen demand to outstrip delivery (e.g. severe sepsis, malignant hyperpyrexia)
- as a specific treatment for certain conditions, e.g. carbon monoxide poisoning, postoperative nausea and vomiting
- to allow humans to survive at very low atmospheric pressure, e.g. mountaineering and high altitude flying.

Oxygen (O_2) is also administered in combination with other gasses for therapeutic or other purposes. For example:

- Entonox
  A mixture of 50% O_2 and 50% nitrous oxide (N_2O) that is used in anaesthesia, as an analgesic both in labour and for short procedures such as reduction of fractures, removal of drains or dressing changes
- Heliox
  A mixture of 21% O_2 and 79% helium that can be used with additional oxygen entrainment in the acute treatment of obstructive pulmonary disorders, e.g. asthma[1]
- Diving gasses (e.g. Trimix)
  Mixtures of oxygen, helium and nitrogen at a variety of different ratios are used to allow deep sea diving.

The method of administration of supplemental oxygen depends on the cause and severity of hypoxaemia. If supplemental oxygen at or just above atmospheric pressure is required to saturate haemoglobin in the bloodstream, the treatment is normally referred to as *normobaric oxygen therapy*. If supplemental oxygen has to be delivered by dissolving it in plasma at pressures greater than atmospheric, it is classified as *hyperbaric oxygen therapy*.

NORMOBARIC OXYGEN THERAPY

There are almost as many devices that deliver oxygen as there are indications for its use. These devices may be classified by the extent to which the patient relies upon them to correct any deficiency in oxygen delivery (Table 7.1).

Low dependency systems

When supplemental oxygen alone is required, it may be delivered by many different devices. By convention these are divided into *variable* and *fixed performance* devices.

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Table 7.1 A classification of oxygen delivery devices by degrees of dependency

	DEFINITION	RESPIRATORY PATTERN	EXAMPLES
Low dependency	When supplemental oxygen alone is sufficient to correct hypoxia	Spontaneous breathing	Nasal prongs, standard facemasks
Medium dependency	When supplemental oxygen and a degree of respiratory assistance is required	Spontaneous breathing but requires additional support, for example CPAP	CPAP mask and equipment
High dependency	When supplemental oxygen and full respiratory support is required	Unreliable or absent. Requires NIPPV or IPPV	Involves use of intensive care, operating theatre or non-invasive ventilators

CPAP, continuous positive airway pressure; NIPPV, non-invasive positive pressure ventilation; IPPV, intermittent positive pressure ventilation



Figure 7.1 Standard adult facemask.

Table 7.2 The major determinants of F_IO_2 delivered by variable performance masks

Equipment factors	Oxygen flow rate Mask volume Quality of mask fit Area of holes in mask
Patient factors	Respiratory rate Tidal volume Peak inspiratory flow rate
Additional factors	Presence of other gasses or vapours, e.g. humidifier

Variable performance devices

With these devices, the oxygen concentration delivered to the upper airway varies with the phase and pattern of respiration (as well as the selected oxygen flow rate). As an example, consider a standard oxygen mask applied to the face (Fig. 7.1). Oxygen flows into the facemask from a continuous supply and quickly fills the relatively small volume of the mask (approximately 200 ml) at the end of an exhaled breath. The oxygen then begins to escape through the mask's vents and where the seal against the face is imperfect.

When the subject breathes in, the oxygen-rich mask contents are inhaled first. If the tidal volume exceeds mask volume, air is then entrained from outside and mixes with the supplied oxygen before entering the subject's upper airway. The fraction of inspired oxygen (F_IO_2) is reduced during this phase as the air dilutes the oxygen, the extent of which depends on the difference between the flow rate of supplied oxygen to the inspiratory flow rate of the patient. In physiological terms, this means that for a given oxygen flow rate, the F_IO_2 will be higher when the patient's respiratory rate, tidal volume and peak inspiratory flow rate are low, and *vice versa*.

The concept of how the difference between peak inspiratory flow rate and oxygen delivery rate determines F_IO_2 is fundamental to the understanding of the performance of oxygen delivery devices and is a recurring theme throughout this chapter.

Mask construction and oxygen flow rate are also important in determining F_IO_2. If mask volume exceeds the patient's tidal volume, the F_IO_2 will be high as there is minimal air entrainment, although carbon dioxide accumulation and rebreathing will occur, especially at low oxygen flows. Masks with volumes substantially smaller than tidal volume tend to collapse during inhalation, making patients feel claustrophobic and thus likely to remove the mask. The ideal mask volume for an adult is approximately 200 ml with vents in the body through which air is entrained if peak inspiratory flow rate is high. Another determinant of F_IO_2 is the rate at which oxygen is supplied into the mask; higher flow rates will result in less air entrainment and hence higher F_IO_2.

The complex interplay between the factors that determine F_IO_2 (Table 7.2) means that most oxygen delivery devices deliver a variable F_IO_2. The extent of this variability

is difficult to quantify as normal methods of measuring breathing pattern are impractical in clinical practice. The latter require a mouthpiece that interferes with oxygen flow in the mask, and subjects are not able to breathe with a fixed respiratory pattern long enough for accurate measurements to be recorded. These difficulties explain why the factors most influential in determining F_IO_2 are still unclear.

As the volume of the mask as a reservoir plays an important role in its performance, low-dependency oxygen delivery devices can be further classified by their reservoir *capacity*.

No capacity oxygen delivery devices

Traditionally, nasal cannulae (also known as nasal prongs or specs) deliver unhumidified oxygen to the nasopharynx (Fig. 7.2). They are more comfortable and less claustrophobic than facemasks, allowing talking, eating and drinking, and making them the most suitable means of delivering long-term oxygen therapy for chronic conditions, especially at home. The main drawback is that an unhumidified oxygen flow greater than 2 L min^{-1} can cause discomfort and drying of the nasal mucosa. Thus unhumidified oxygen therapy in association with the tiny capacity of the nasal cannulae leads to variable F_IO_2. Clinical and laboratory studies in this context have shown an enormous variability with respiratory pattern, F_IO_2 ranging between 0.26 and 0.90.[2,3] A further design is the nasal sponge (tipped) catheter, which is lodged in one nostril and again effectively uses the nasopharynx as a small reservoir during the respiratory pause (Fig. 7.3). Devices permitting humidified oxygen therapy via nasal cannulae are now available (Fig. 7.4), allowing oxygen flow rates up to 60 L min^{-1} to be tolerated for long periods. As a result, consistently higher, more reliable F_IO_2 can be achieved with this form of high flow humidified oxygen therapy.[4] However, warm humidified gas is a good medium for bacterial growth; therefore, it is mandatory that any disinfection and infection control procedures deemed necessary by manufacturers and regulatory authorities are adhered to. The other point to note is that delivering high flows into the nasopharynx generates a small amount of positive end expiratory pressure (PEEP) depending on the flow used and if the mouth is open or closed. Pharyngeal pressures of up to 4 cm H_2O have been recorded with a flow of 60 L min^{-1} with the mouth open, increasing to nearly 10 cm H_2O with the mouth closed.[5] As such, high flow humidified oxygen therapy delivered via nasal cannulae should be regarded as a *medium dependency device* (see below). Its role in the management of acutely breathless patients is becoming increasingly appreciated.



Figure 7.2 Nasal cannulae.

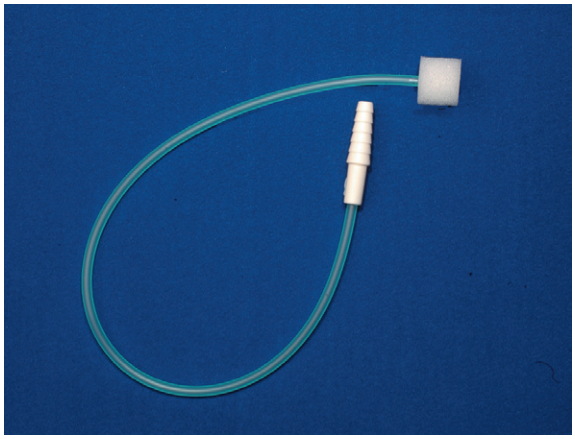


Figure 7.3 Nasal sponge tipped catheter.



Figure 7.4 Nasal cannulae for supplying humidified oxygen. *With permission of Fisher & Paykel Healthcare.*



Figure 7.7 High-capacity oxygen delivery device: Facemask with reservoir bag. Note expiratory flap valve visible on body of mask.



Figure 7.8 Hi-Ox Mask, Viasys Healthcare. The blue tube is an additional nebulisation port.

reservoir bag but designed such that it also allows entrainment of room air if the bag is emptied (Fig. 7.9). This and the expiratory one way valve are neatly arranged within the front of the mask itself. Additionally this device is made of non-PVC materials, in an attempt to reduce the impact on the environment during disposal.

Generally, it is accepted that facemasks with reservoir bags probably deliver an F_IO_2 between 0.6 and 0.8;[9,10] however, in reality there is likely to be considerable variation around these values due to leak between the mask and face combined with variable respiratory patterns. Best results will be achieved by using an O_2 flow rate adequate for the patient's needs such that the reservoir bag empties by no more than a third during inspiration and by achieving the best seal possible between the mask and the face. Some masks incorporate additional features, for example a small chimney containing a red polystyrene ball that moves up and down with respiration so that a patient's respiratory effort can be visualized and respiratory rate can easily be measured (Fig. 7.10).[11]

A reservoir bag may also be attached to airway maintenance devices to improve the F_IO_2 delivered during emergence from anaesthesia. The T-Bag (Ultimate Medical Pty Ltd, Australia) (Fig. 7.11) consists of a reservoir bag with a capacity of 300 ml, attached via a 15 mm internal diameter ISO connector to a laryngeal mask (LM) or endotracheal tube (TT). There is a 3 mm connector that feeds oxygen directly into the reservoir bag and also a 10 mm port, which is open to the atmosphere. During inspiration, the reservoir bag, with its 15 mm aperture, provides a greater proportion of inspiratory gas. A smaller amount is entrained through the 10 mm port, which has a higher resistance to flow. During exhalation, this process is reversed until the reservoir bag is full. Subsequent exhalate then leaves via the 10 mm port only. The T-Bag is light enough not to displace the LM, provides visual confirmation of spontaneous ventilation and supplies a F_IO_2 of up to 0.70 with an oxygen supply rate of 6 l min^{-1}. It is also possible to ventilate patients briefly by occluding the 10 mm port with a digit, with additional volume being delivered by squeezing the bag.

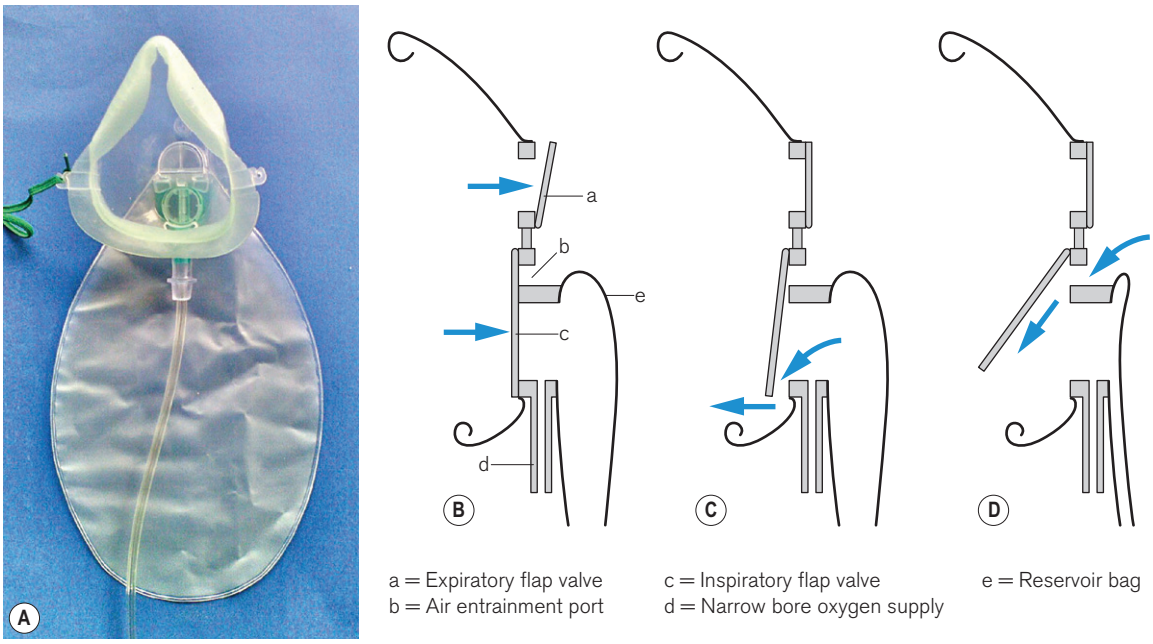


Figure 7.9 **A.** High-capacity mask (Intersurgical, UK). **B.** Position of valves in exhalation. **C.** In inhalation with oxygen flowing. **D.** In inhalation in the event of oxygen supply failure.



Figure 7.10 A Respi-Check facemask with device for detecting respiratory effort.

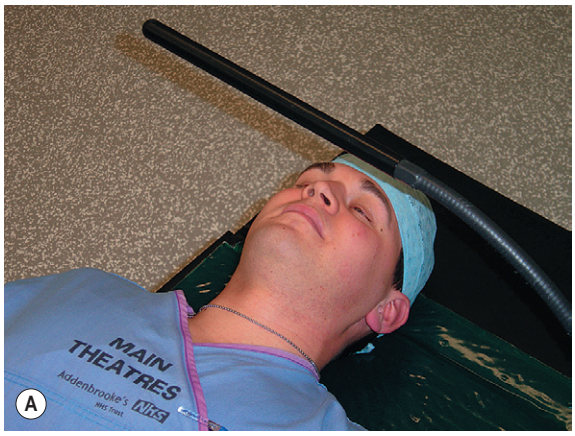


Figure 7.11 The T-Bag device for delivering supplemental oxygen via laryngeal mask airway or endotracheal tube.

Very high capacity oxygen delivery devices (capacity >2500 ml)

Babies tolerate facemasks poorly and whilst small nasal cannulae are available, oxygen is frequently delivered in an incubator or via a head box or tent. Incubators make use of a metered oxygen source (see below) to allow F_IO_2 to be chosen with precision.

During ophthalmic surgery under local anaesthesia, oxygen is sometimes administered under the drapes using an eye surgery bar (Fig. 7.12A). The bar is hollow (Fig. 7.12B). Oxygen is fed into the bar at one end and escapes through perforations in the bar where it is in close proximity to the patient. There are many different designs



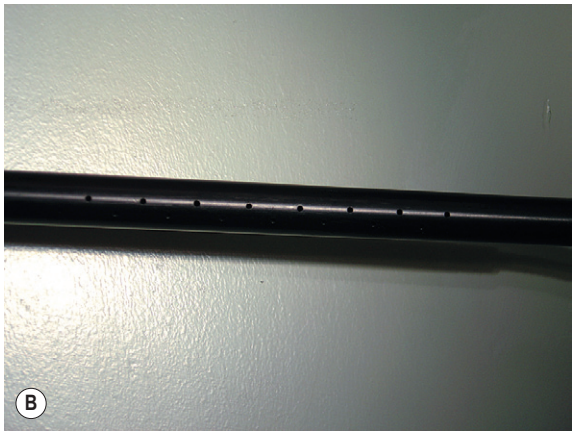


Figure 7.12 **A.** Oxygen bar used for patients undergoing eye surgery under local anaesthesia. **B.** Note the holes in the bar for the administration of oxygen.

and some are 'home made'. Oxygen flow under the drapes (which acts as a large reservoir) increases the F_IO_2, helps reduce rebreathing of CO_2, and provides a refreshing breeze on the patient's face, when otherwise they may begin to feel hot and claustrophobic.[12]

Fixed performance devices

When supplemental oxygen is delivered, the exact F_IO_2 is often irrelevant, as long as inhaled air is enriched with sufficient additional oxygen so that oxygen delivery to the tissues can meet their oxygen demand. However, there are circumstances when it is important to deliver a known F_IO_2 that does not vary with respiratory pattern. These are:

- when following guidelines for the management of patients with exacerbation of COPD, which state that a fixed F_IO_2 be employed[13]
- when administering supplemental oxygen to a critically ill patient for whom an accurate calculation of the P_aO_2 / F_IO_2 ratio is needed to guide decisions about intubation or diagnose acute respiratory distress syndrome.[14]

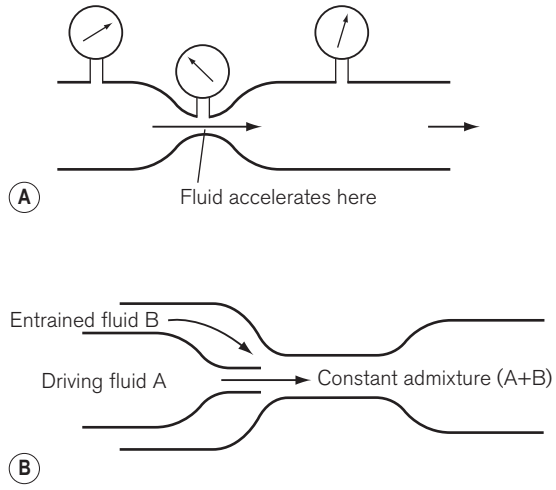


Figure 7.13 **A.** Bernoulli effect. **B.** A Venturi.

Under these circumstances a fixed performance device will provide an F_IO_2 less dependent on respiratory pattern. These devices make use of the Venturi principle.

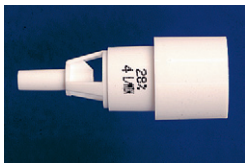
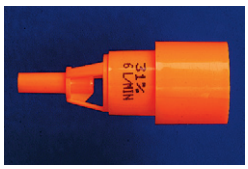
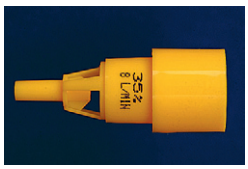
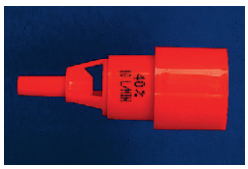
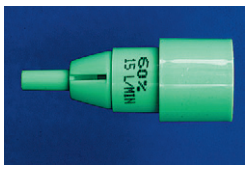
The Bernoulli effect and the Venturi principle

The Bernoulli effect (Fig. 7.13A) describes the change in pressure that occurs when a fluid flows through a constriction. The fluid accelerates, gaining kinetic energy at the expense of potential energy; as a result, the pressure distal to the constriction is reduced. The Venturi principle uses this phenomenon to allow a second fluid to be entrained into the stream of the first, either through a side arm that opens into the area of low pressure or via a co-axial arrangement (Fig. 7.13B).

Referring to Fig. 7.13B, the degree of pressure drop and resultant entrainment is dependent on the flow rate of fluid A, the physical dimensions of the constriction and the density and viscosity of fluids A and B. Hence Venturis for oxygen delivery devices can be designed to entrain a known amount of air into a stream of oxygen at a specified flow rate, providing a fixed final F_IO_2 (Table 7.3).

The performance of Venturi devices is less dependent on respiratory pattern as they use relatively high oxygen flow rates to which is added the flow rate of entrained air. The total flow rate is close to or exceeds the patient's peak inspiratory flow rate, and is sufficient to flush away alveolar gas expired into the mask. The Venturi devices that entrain most air deliver a lower F_IO_2 and are thus most reliable, but those that deliver high F_IO_2, especially the 60% O_2 device, entrain less air and also have a lower total flow rate into the mask, which may be inadequate in a patient breathing with a high peak inspiratory flow rate

Table 7.3 Fraction of inspired oxygen, oxygen flow rates and air entrainment of widely used Venturi devices

F_iO_2 BY VENTURI VALVE	NOZZLE COLOUR	OXYGEN FLOW RATE TO VENTURI VALVE (l min^{-1})	AMOUNT OF AIR ENTRAINED (l min^{-1})	TOTAL FLOW TO PATIENT (l min^{-1})
0.24		2	51	53
0.28		4	41	45
0.31		6	41	47
0.35		8	37	45
0.40		10	32	42
0.60		15	15	30

and can also result in rebreathing. Hence the highest F_iO_2 Venturi devices may underperform by 5–10%.[15]

Numerous manufacturers produce single-use clear plastic oxygen masks which come prepacked with a number of different Venturi injectors (Fig. 7.14), typically giving overall flow rates of 30–60 l min^{-1}, as detailed in Table 7.3. Adjustable Venturi devices are also available (Fig. 7.15), where the user can select the F_iO_2 required by adjusting the size of the air entrainment aperture, before the total fresh gas flow is passed through a humidifier.

A Venturi can also be attached to a T-piece (Fig. 7.16) so that supplemental oxygen can be delivered to patients who are breathing spontaneously through a supraglottic airway or less commonly a tracheal tube.[16] These devices are used on a short-term basis in recovery areas during emergence from anaesthesia. The segment of 22 mm diameter corrugated tubing has a volume of 56 ml, a



Figure 7.17 CPAP equipment including humidifier.



Figure 7.18 CaStar non-invasive ventilation/CPAP hood. StarMed SpA, Italy. (Photograph courtesy of StarMed SpA).

The fresh gas flow in the CPAP circuit must exceed maximal inspiratory gas flow in order to maintain the same positive pressure throughout the respiratory cycle. Some systems incorporate a spring loaded reservoir bag; otherwise a flow generator connected to the piped high pressure oxygen outlet ('wall' CPAP) is used (Fig. 7.20). The flow generator is a high-pressure oxygen-driven Venturi injector



Figure 7.19 7.5 cm H_2O CPAP valve.

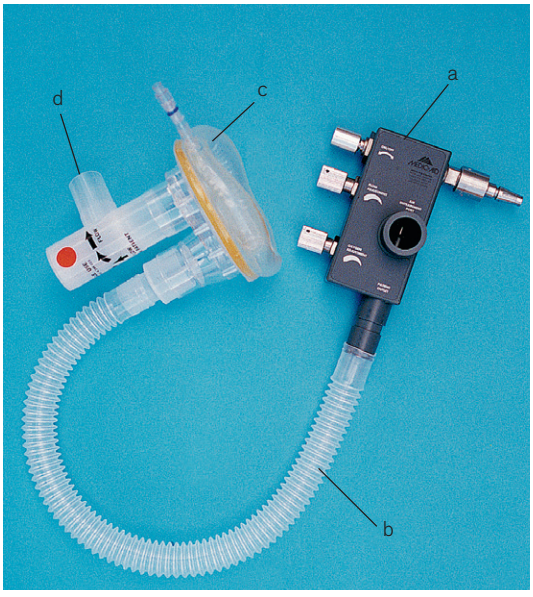


Figure 7.20 A CPAP device. The flow generator **(a),** plugs into a terminal outlet for oxygen. It has an ON switch, a flow adjustment and an oxygen concentration adjustment. A section of wide-bore tubing **(b),** connects it to a close-fitting facemask **(c),** that has two one-way valves in the body of the mask. One is an inlet valve and the other, fitted with a disposable CPAP valve **(d),** is the outlet valve.

and most devices of this nature tend to be quite noisy. To ensure that CPAP is maintained during all phases of respiration, it is important to check that the CPAP valve is being kept open by an adequate flow throughout the respiratory cycle. Humidification and oxygen monitoring units can be added to the system (Fig. 7.17). Humidification is difficult, however, when using a CPAP helmet due to problematic steaming and condensation.

Intubated patients being weaned from a ventilator may receive oxygen and CPAP through a T-piece arrangement with a CPAP valve connected to the distal end. This has

the advantage of doing away with the work of breathing necessary for triggering gas flow from a ventilator.

High-dependency systems

In addition to supplying supplemental oxygen, brief periods of low-level positive pressure ventilation may be applied by means of a nasal mask or facemask, avoiding the need for tracheal intubation. Non-invasive positive pressure ventilation (NIPPV or 'NIPPY') can be used electively to treat patients with central apnoea syndromes, neuromuscular or chest wall disease. It is increasingly used in acute respiratory failure and weaning.[20] The equipment used is the same as that required for CPAP, with the addition of a sophisticated intensive care ventilator capable of providing synchronized intermittent mandatory bi-level ventilation. For home use, highly sophisticated CPAP and BiPAP equipment is available, including some that use microprocessor control to adjust their own settings and can be controlled by telemetry.[21]

Metered sources of oxygen and air

The high flow rates required for CPAP need a metered source of fresh gas. The flow generator in the CPAP equipment (Fig. 7.20) plugs into a terminal outlet for oxygen via a Schrader probe (a quick release, gas specific connector, Chapter 1), typically the hospital piped oxygen supply or, sometimes, an oxygen cylinder. It has an 'on' switch and adjustments for F_IO_2 and flow rate. Air is entrained into the device through an adjustable Venturi orifice to determine F_IO_2 and flow rate is adjusted by means of a variable pressure regulator. Flow generators such as these can also be used in conjunction with a humidifier to provide high flow oxygen therapy delivered through a conventional loose fitting mask (aerosol mask) or high flow nasal cannulae systems (Fig. 7.4).

THE ADMINISTRATION OF OXYGEN IN A MIXTURE OF GASSES

Entonox

The physical properties, storage and supply of this 50:50 mixture of oxygen and nitrous oxide (N_2O) are described in Chapter 1. This chapter will describe the equipment used for the self-administration of Entonox during labour and for short painful procedures such as fracture reduction, drain removal or dressing changes.

Entonox is usually delivered by means of a demand valve connected to either a facemask or mouthpiece. These devices are easy to use, allowing patients to breathe Entonox with minimal supervision and instruction from paramedics, midwives or nursing staff. However, staff must be aware that Entonox cylinders should not be stored in areas where temperature is liable to fall below 0°C. The pseudocritical temperature of Entonox is pressure dependent and may be as high as –5.5°C, at which N_2O separates out of the gaseous mixture and enters the liquid phase by lamination. The first breaths from such a cold cylinder are predominantly oxygen, but as the cylinder empties a hypoxic mixture may be delivered.

The BOC Entonox valve

The original valve (Fig. 7.21) is no longer manufactured, but is described here as many are still in use. It clamps directly to a pin-index Entonox cylinder and is composed of a first-stage pressure regulator and second-stage demand valve. The demand valve consists of a sensitive rubber diaphragm that is deformed by negative pressure generated when a patient begins to inhale. The diaphragm operates a push rod that in turn tilts a flap to open the valve. A corrugated hose carries the Entonox to a facemask or mouthpiece, which contains an exhalation valve. The sensing diaphragm can be manually depressed to test the integrity of the system.

The successor to this valve is shown in figure 7.22. Note that like the Pneupac valve below it relies on a separate and remote first stage pressure reduction valve to deliver gasses at 4 bar to the valve.

The Pneupac Entonox valve

This valve is shown in figure 7.23 connected to a lightweight Entonox cylinder with an integral pressure reducing valve. The advantage of this system compared to the original BOC valve in figure 7.21 is that the high pressure hose may be several metres long so that the Entonox cylinder can be stored remotely from the facemask. This allows the cylinder to be kept warm inside the ambulance when administering Entonox at the roadside in cold conditions. This model also incorporates a 'push to test' feature that can be used for manual insufflation if desired.

Heliox

Heliox is presented in the UK as Heliox21 – a mixture of 21% oxygen and 79% helium. Helium is less dense than oxygen and nitrogen, and hence is more likely to flow in a laminar fashion in narrow airways, reducing the amount of less efficient turbulent flow (and the majority of transitional flow states). Additionally, this allows faster diffusion of oxygen and carbon dioxide in the distal parts of the lung. Published literature describes the use of Heliox in acute asthma, chronic obstructive pulmonary disease (COPD) exacerbation, croup, bronchiolitis, post extubation stridor, as well as upper airway obstruction.[22] Whilst there is little doubt of its usefulness in upper airway obstruction, the evidence for its use in lower airway obstruction is less convincing, leading Cochrane reviewers to conclude that Heliox therapy has little to offer in the

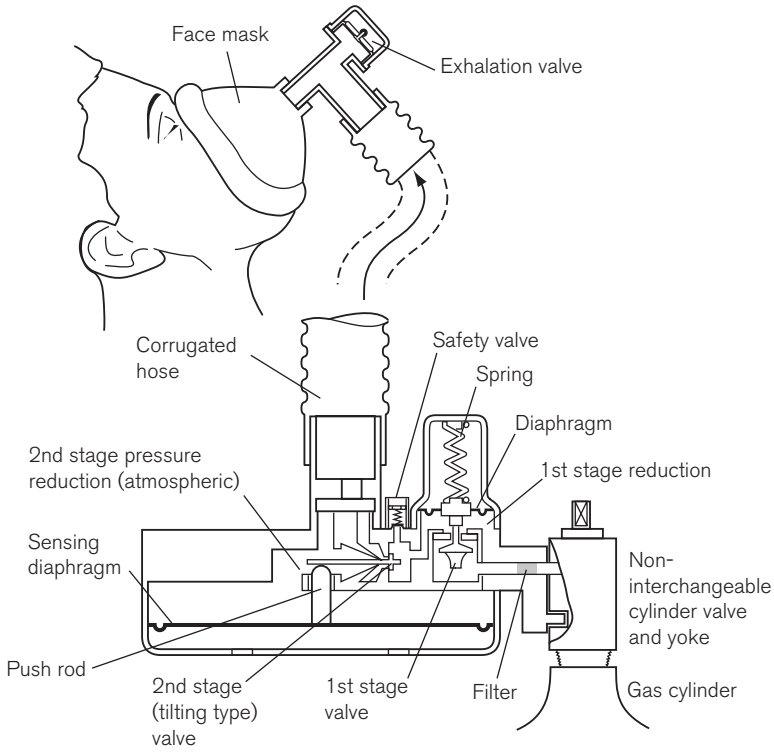


Figure 7.21 Working principles of the Entonox valve.



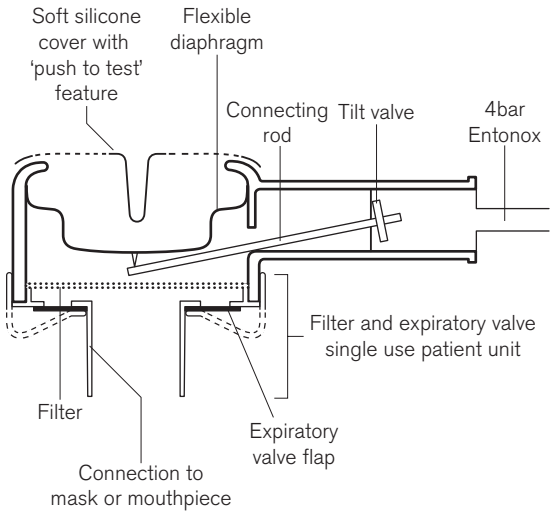


Figure 7.22 Carnét Analgesic Demand Valve (BPR Medical, UK) together with working principles. A single use patient unit comprising filter, expiratory flap valve and 22 mm ISO connection to mouthpiece or facemask, simplifies infection control issues.
Photograph courtesy of Richard Radford at BPR Medical.



Figure 7.25 BOC Helontix Breathing Kit.

may allow for better drug delivery to the lung periphery due to increased flows through the upper generations of the bronchial tree. However, Heliox reduces the efficiency of the nebulizer and so increased flows are required (compared to oxygen or air nebulizer flows).[24] Because of the need to rigidly exclude room air from the inhaled Heliox mixture, it is best to use a gas-tight delivery system with an adequate reservoir combined with a leak-free interface to the patient's face.[7,24] This type of breathing system must incorporate an entrainment valve, so that the patient can breathe room air in the event of fresh gas flow failure. Humidification of Heliox may also be important for longer-term use. A recently introduced breathing system (BOC Medical) aims to meet the needs of a specific Heliox administration system. The device consists of a 2L reservoir bag attached to the inlet of an angle piece that houses a unidirectional inspiratory valve, the entrainment valve, an expiratory valve and an inlet for the Heliox. The outlet has a 22 mm male ISO connector that can be attached to an anaesthetic or CPAP/BiPAP facemask (Fig. 7.25).

Heliox is sometimes administered to ventilated patients, especially children, via the air inlet of a ventilator. The altered density of the fresh gas within the ventilator has a significant and unpredictable effect on delivered F_IO_2 and tidal volume.[25] Extra care should be taken in these circumstances to avoid barotrauma and detect over- or under-ventilation.

Heliox compatible anaesthetic machines and ventilators, able to measure gas flows containing helium mixtures are gradually becoming more widely available (e.g. the AVEA ventilator, CareFusion, UK). By facilitating administration, this should in due course allow a more defined role for Heliox therapy to be characterized in both non-invasively and invasively ventilated patients.

OXYGEN DELIVERY AT HIGH OR LOW ATMOSPHERIC PRESSURES

Hyperbaric medicine

Hyperbaric oxygen is most commonly used to treat decompression sickness and carbon monoxide poisoning.[26] Breathing oxygen at three times normal atmospheric pressure (absolute pressure) increases the amount of dissolved oxygen in plasma from a maximum of 2 ml 100 ml^{-1} to 6 ml 100 ml^{-1}, resulting in an extra 200 ml min^{-1} of oxygen delivery. As a result of this perceived physiological benefit, the US Food and Drug Administration (FDA) has approved 11 other application areas for hyperbaric oxygen, including the treatment of gas gangrene, necrotizing fasciitis, osteoradionecrosis, skin flaps and grafts, chronic refractory osteomyelitis, thermal burns and compartment syndrome.[27] However, the evidence for its efficacy in these latter conditions is unclear; most have not been investigated in clinical trials and many are based upon case reports or small series.

There are three types of hyperbaric chamber. Those in medical use tend to be divided into type A and type B on the basis of their size. The type A chamber (Fig. 7.26A and B) is large enough to accommodate one or more critically ill ventilated patients plus medical staff and equipment. The chamber shown was situated in the Hyperbaric Medicine Unit in the Royal Hospital, Haslar, UK. It could be adapted to accommodate ten or more sitting patients. Large chambers may have air conditioning and humidity control systems as the temperature of air rises as it is compressed. Hyperbaric chambers have multiple antistatic points to prevent build-up of electrical charge, an essential safety feature, as a spark in an enriched oxygen environment at 3 atm would result in a violent explosion. Compressing the chamber with air whilst the occupants breathe oxygen via a tightly fitting facemask (Fig. 7.26C) or head tent further reduces risk of explosion.

Type B chambers are smaller and less suitable for ventilated patients. In the UK, hyperbaric chambers are often situated in deep sea diving centres, although small, single-patient portable chambers are also available (Fig. 7.26D). Lightweight and portable equipment has been developed to treat acute life-threatening mountain sickness and decompression sickness on the scene, to transport patients suffering from decompression sickness and to pressurize patients during flight in non-pressurized aeroplanes.[28] These are sometimes referred to as '*mild chambers*' in reference to the much lower pressures achieved (1.3 atm absolute) with these soft-sided inflatable chambers (Fig. 7.26E).

Diving

Divers face several problems when breathing underwater at increased atmospheric pressure.[29] Pressure increases by one atmosphere for every 10 m of descent:

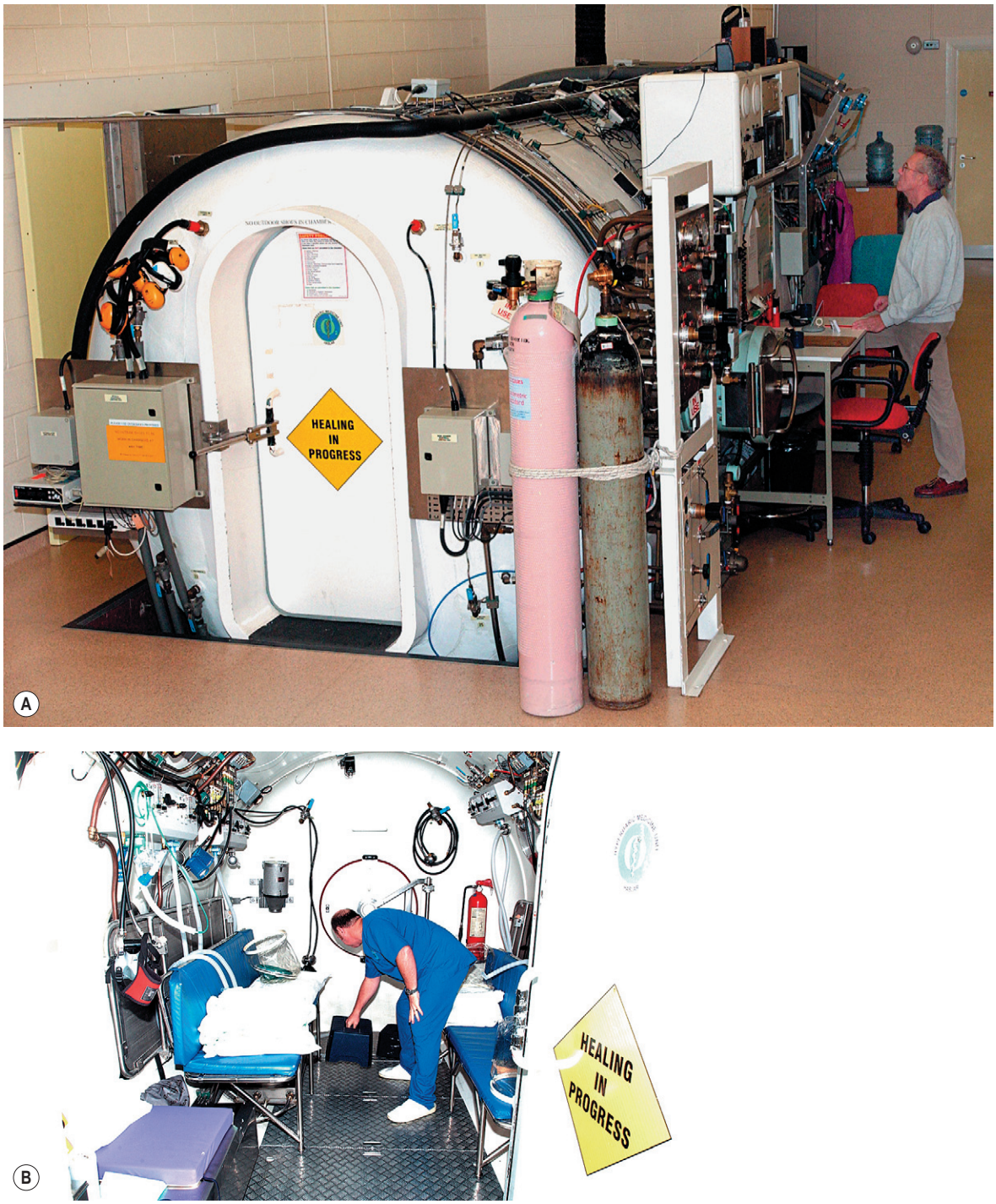


Figure 7.26 **A.** and **B.** The Type A hyperbaric chamber at the former Royal Hospital, Haslar (courtesy of Royal Navy and Qinetiq Ltd).

Continued







Figure 7.26 cont'd. **C.** Single-patient portable hyperbaric chamber and **D.** tightly fitting oxygen mask for use inside the chamber (courtesy of HYOX Ltd). **E.** Soft-sided hyperbaric chamber. Photo courtesy Bruce McKeeman, Summit To Sea.

- Nitrogen: this poorly soluble gas is thought to be physiologically inert, but at high pressure it is forced into solution in the tissues. If ascent is too rapid, bubbles of N_2 form causing decompression sickness ('the bends'), which is potentially fatal. At depths of 50 m or more, N_2 causes narcosis by an unknown mechanism (*rapture of the deep*).
- Oxygen: high concentrations of oxygen cause lung injury and are toxic to the central nervous system. Convulsions can occur after breathing O_2 at 4 atm for 30 min.

To combat these problems divers use a variety of gas mixtures: to increase the amount of time that they may stay at depth without the need for decompression stops when ascending, to reduce the risk of decompression sickness by decreasing the amount of dissolved nitrogen in the body and to avoid oxygen toxicity. Nitrogen may be partially or completely replaced by helium, which is not narcotic and reduces the work of breathing at very high pressures. For very deep dives, the oxygen content of the diving gas may be as low as 1%.[30]

Most divers use an open-circuit *s*elf-*c*ontained *u*nderwater *b*reathing *a*pparatus (Scuba).[30] The chosen gas is contained in steel, aluminium or titanium cylinders. Tank volume is normally about 10 L, but may be very small ('bail out bottles') or up to 15 L for deeper or longer exposures. A balanced first-stage regulator situated on the tank reduces gas pressure to 10 atm. Gas then passes through an intermediate hose to a second regulator on the mouthpiece where it is balanced to the pressure of the surroundings and the lungs. When the diver exerts a slight negative pressure on the regulator at the start of inspiration, a non-return valve between the hose and mouthpiece opens and initiates fresh gas flow. Expiration and positive pressure closes the valve and expired gas is lost to the open water. Scuba equipment allows easy breathing with little resistance to inhalation and expiration independent of tank pressure. Safety is of paramount importance: tanks are visually inspected annually and subjected to a hydrostatic test procedure every 5 years and the resistance offered by each regulator, accuracy of instruments and integrity of hoses are checked regularly. An additional safety feature is a pinhole orifice in the proximal end of the high-pressure hose, which prevents injury from a flailing hose should it rupture.

Military, cave and specialist divers may choose to use a closed circuit system in shallow dives when it is important not to produce bubbles.[30] The diver typically breathes 100% oxygen via a reservoir bag, which is then exhaled via a non-return valve into a canister containing a CO_2 absorber. The expired oxygen is recycled back into the reservoir bag, which is supplemented by additional oxygen

from the high-pressure supply. Use of mixed gasses may allow deeper dives using a closed circuit system, with electronic monitoring of gas levels to ensure that their levels are maintained within safe ranges. A high degree of training and expertise is required to dive safely with such equipment. This system offers a higher resistance to breathing and CO_2 is liable to accumulate due to absorber inefficiency in moist, cold environments.

Mountaineering

Mountaineers require oxygen to facilitate climbing at very high altitude or to manage medical emergencies.[31] Closed circuit systems that use soda lime to absorb expired carbon dioxide have been used, but are unreliable, as internal valves tend to ice up before the soda lime begins to warm up and work. Constant-flow systems have become most popular, mainly because of their simplicity. Weighing 3–7 kg in total, they comprise light titanium cylinders containing special no water oxygen at 306 atm, pressure reducing valves and regulators. The regulator permits flow of 0.5–4 l min^{-1} into a rubber or plastic reservoir and facemask via light, non-kinking tubing. Finally, the mask is secured to the face in a helmet or with straps. All systems are checked and tested for several hours in a 'cold room' at −20 °C before an expedition, paying particular attention to humidification, and the pooling of condensate or deposition of ice particles.[32]

REFERENCES

1. Ho AM, Lee A, Karmakar MK, Dion PW, Chung DC, Contardi LH. Heliox vs air-oxygen mixtures for the treatment of patients with acute asthma: a systematic overview. *Chest* 2003;**123**:882–90.
2. Collis JM, Bethune DW. Oxygen by face mask and nasal catheter. *Lancet* 1967;**1**:787–8.
3. Ooi R, Joshi P, Soni N. An evaluation of oxygen delivery using nasal prongs. *Anaesthesia* 1992;**47**:591–3.
4. Sim MA, Dean P, Kinsella J, Black R, Carter R, Hughes M. Performance of oxygen delivery devices when the breathing pattern of respiratory failure is simulated. *Anaesthesia* 2008;**63**:938–40.
5. Groves N, Tobin A. High flow nasal oxygen generates positive airway pressure in adult volunteers. *Aust Crit Care* 2007;**20**:126–31.
6. Campkin NT, Ooi RG, Soni NC. The rebreathing characteristics of the Hudson oxygen mask. *Anaesthesia* 1993;**48**:239–42.
7. Standley TD, Smith HL, Brennan LJ, Wilkins IA, Bradley PG, Barrera Groba C, et al. Room air dilution of Heliox given by facemask. *Intensive Care Med* 2008;**34**:1469–76.
8. Slessarev M, Somogyi R, Preiss D, Vesely A, Sasano H, Fisher JA. Efficiency of oxygen administration: sequential gas delivery versus 'flow into a cone' methods. *Crit Care Med* 2006;**34**:829–34.
9. Cairo JM. Administering medical gasses: regulators, flowmeters, and controlling devices. In: Cairo JM, Pilbeam SP, editors. *Mosby's respiratory care equipment*. St. Louis: Mosby; 1999. p. 62–88.
10. Branson RD. The nuts and bolts of increasing arterial oxygenation: devices and techniques. *Respir Care* 1993;**38**:672–86; discussion 87–9.
11. Breakell A, Townsend-Rose C. The clinical evaluation of the Respi-check mask: a new oxygen mask incorporating a breathing indicator. *Emerg Med J* 2001;**18**: 366–9.
12. Rayen AT. A device for oxygen administration during ophthalmic surgery under local anaesthesia. *Anaesthesia* 2000;**55**:508–9.
13. O'Driscoll BR, Howard LS, Davison AG. BTS guideline for emergency oxygen use in adult patients. *Thorax* 2008;**63**(Suppl 6): vi1–68.
14. Bernard GR, Artigas A, Brigham KL, Carlet J, Falke K, Hudson L, et al. The American-European Consensus Conference on ARDS. Definitions, mechanisms, relevant outcomes, and clinical trial coordination. *Am J Respir Crit Care Med* 1994;**149**:818–24.
15. Jones HA, Turner SL, Hughes JM. Performance of the large-reservoir oxygen mask (Ventimask). *Lancet* 1984;**1**:1427–31.
16. Campbell DJ, Fairfield MC. The delivery of oxygen by a Venturi T piece. *Anaesthesia* 1996;**51**: 558–60.
17. Duncan AW, Oh TE, Hillman DR. PEEP and CPAP. *Anaesth Intensive Care* 1986;**14**:236–50.
18. Lindner KH, Lotz P, Ahnefeld FW. Continuous positive airway pressure effect on functional residual capacity, vital capacity and its subdivisions. *Chest* 1987;**92**: 66–70.
19. Nava S, Hill N. Non-invasive ventilation in acute respiratory failure. *Lancet* 2009;**374**:250–9.
20. Brochard L, Mancebo J, Elliott MW. Noninvasive ventilation for acute respiratory failure. *Eur Respir J* 2002;**19**:712–21.
21. Highcock MP, Morrish E, Jamieson S, Shneerson JM, Smith IE. An overnight comparison of two ventilators used in the treatment of chronic respiratory failure. *Eur Respir J* 2002;**20**: 942–5.
22. Ball JAS, Rhodes A, Grounds RM. A review of the use of helium in the treatment of acute respiratory failure. *Clin Intensive Care* 2001;**12**: 105–13.
23. Rodrigo G, Pollack C, Rodrigo C, Rowe BH. Heliox for nonintubated acute asthma patients. Cochrane Database Systematic Review 2006;4:CD002884.
24. Corcoran TE, Gamard S. Development of aerosol drug delivery with helium oxygen gas mixtures. *J Aerosol Med* 2004;**17**: 299–309.
25. Berkenbosch JW, Grueber RE, Dabbagh O, McKibben AW. Effect

of helium-oxygen (Heliox) gas mixtures on the function of four pediatric ventilators. *Crit Care Med* 2003;**31**:2052–8.

26. Tibbles PM, Perrotta PL. Treatment of carbon monoxide poisoning: a critical review of human outcome studies comparing normobaric oxygen with hyperbaric oxygen. *Ann Emerg Med* 1994;**24**:269–76.
27. Guo S, Counte MA, Romeis JC. Hyperbaric oxygen technology: an overview of its applications, efficacy, and cost-effectiveness. *Int J Technol Assess Health Care* 2003;**19**: 339–46.
28. Dubois C, Herry JP, Kayser B. Portable hyperbaric medicine, some history. *J Wilderness Med* 1994;**5**:190–8.
29. Spira A. Diving and marine medicine review part I: diving physics and physiology. *J Travel Med* 1999;**6**:32–44.
30. Egstrom GH. Diving equipment. In: Bove AA, editor. *Bove and Davis' Diving medicine*. 3rd ed. London: Saunders; 1997. p. 26–38.
31. West JB. Man at extreme altitude. *J Appl Physiol* 1982;**52**:1393–9.
32. Hendricks DM, Pollock NW, Natoli MJ, Vann RD. Mountaineering oxygen mask performance at 4572 m. *Aviat Space Environ Med* 2000;**71**:1142–7.

Chapter 8

Manual resuscitators

Andrew J Davey

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There are occasions, both in and out of hospital, when a patient needs emergency ventilatory support that requires a device that is easily portable and that does not rely on a source of pressurized gas or electricity for its operation. A manual resuscitator fulfils these requirements. The number of different manufacturers marketing these devices bears testimony to their usefulness. Although there have been a plethora of designs from the first 'Ambu bag' in 1956, they all have three similar components:

1. A self-inflating bag
2. A non-rebreathing valve
3. A facility for the admixture of oxygen.

COMPONENTS

The self-inflating bag

This bag may be made of silicone rubber or polyvinyl chloride (PVC), which is strengthened either by making its wall thicker, or by incorporating circular 'ribs' of identical material during manufacture (Fig. 8.1), so that in the resting state it is expanded. Another design (Ambu Mark IV, Fig. 8.2) has an outer covering of chloroprene or butyl rubber and a thick foam inner layer, which makes the bag expand in the resting state. All devices produced now are latex-free. The PVC versions are designed to be for single use only and the others are made to be steam autoclaved.

The respirable gas inlet mechanism is housed at one end and the non-rebreathing valve at the other.

The respirable gas inlet

This inlet has a number of components (Fig. 8.3):

- *A one-way flap valve* (A). This is fitted to the inlet of the self-inflating bag. When the bag is squeezed, the gas pressure inside the bag rises and causes the flap valve to close. This prevents the escape of gas back through the inlet. When the bag is released, its self-inflating characteristic causes fresh gas from the respirable gas inlet to be indrawn. This may be air, oxygen or a mixture of both.
- *A small bore nipple* (B). This is mounted on the inlet, to allow admixture of oxygen.
- *A wide bore inlet* (C). This supplies the bulk of the gas entering the bag and is usually air, unless oxygen is added, as above. In the latter situation, the final concentration of oxygen delivered is a function of the amount of added oxygen and its dilution with air in the self-inflating bag.
- *A reservoir system* (D). The inlet (C) may be fitted with a reservoir system. This feature is now widely used in almost all manual resuscitators. Its purpose is to store the oxygen fed into the system from the nipple (B). When the minute volume of oxygen supplied is greater than the volume given to the patient, the bag (D) will expand and will provide all the gas for ventilation (i.e. 100% oxygen). The reservoir must be fitted with an overflow valve (E) to prevent overfilling from too high a flow of oxygen and an entrainment valve (F) to allow ingress of air

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Table 8.1 Oxygen concentrations (%) in a Laerdal resuscitator

Adult: Ventilation bag volume 1600 ml; reservoir bag volume 2600 ml

O_2 flow (l min^{-1})	Tidal vol. (ml) × bag cycling rate cycling per min					
	500 × 12	500 × 24	750 × 12	750 × 24	1000 × 12	1000 × 24
3	56 (37)*	39 (32)	47 (33)	34 (29)	41 (32)	30 (28)
5	81 (52)	52 (38)	62 (41)	42 (33)	52 (39)	38 (31)
10	100 (73)	84 (48)	100 (56)	65 (42)	84 (55)	53 (39)
12	100 (84)	97 (53)	100 (61)	74 (45)	94 (60)	59 (42)
15	100 (89)	100 (59)	100 (69)	86 (48)	100 (69)	66 (44)

Data are O_2 concentrations using reservoir (without reservoir)

Child: Ventilation bag volume 500 ml; reservoir bag volume 2600 ml

O_2 flow (l min^{-1})	Tidal vol. (ml) × bag cycling rate per min			
	250 × 20		100 × 30	
	w/reservoir	wo/reservoir	w/reservoir	wo/reservoir
10	100	75	100	90

Infant: Ventilation bag volume 240 ml; reservoir bag volume 600 ml

O_2 flow (l min^{-1})	Tidal vol. (ml) × bag cycling rate per min			
	40 × 30		20 × 40	
	w/reservoir	wo/reservoir	w/reservoir	wo/reservoir
4	98	89	98	98

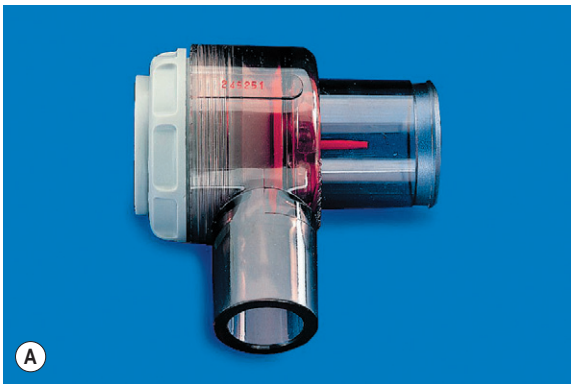
Table 8.2 Oxygen concentrations (%) in the Ambu system with reservoir

O_2 flow (l min^{-1})	Ventilation volume (ml) × frequency			
	250 × 12	600 × 12	750 × 12	1000 × 12
2	74	43	38	34
5	100	76	65	54
10	100	100	100	87
15	100	100	100	100

and occlude the expiratory port. The same movement opens the patient port to allow ingress of the gas.

At the beginning of the exhalation phase, exhaled gas impinges on the convex aspect of the shutter, causing it to move in the opposite direction, so that it opens the expiratory pathway as well as occluding the inspiratory port.

With spontaneous respiration, as inspiratory resistance (0.7 kPa at 10 l min^{-1}) through the valve is less than expiratory resistance (0.8 kPa at 10 l min^{-1}), gas will be drawn preferentially from the bag. Initial movement of gas will also cause the shutter valve to occlude the expiratory path so that the valve behaves in a similar manner to controlled ventilation.





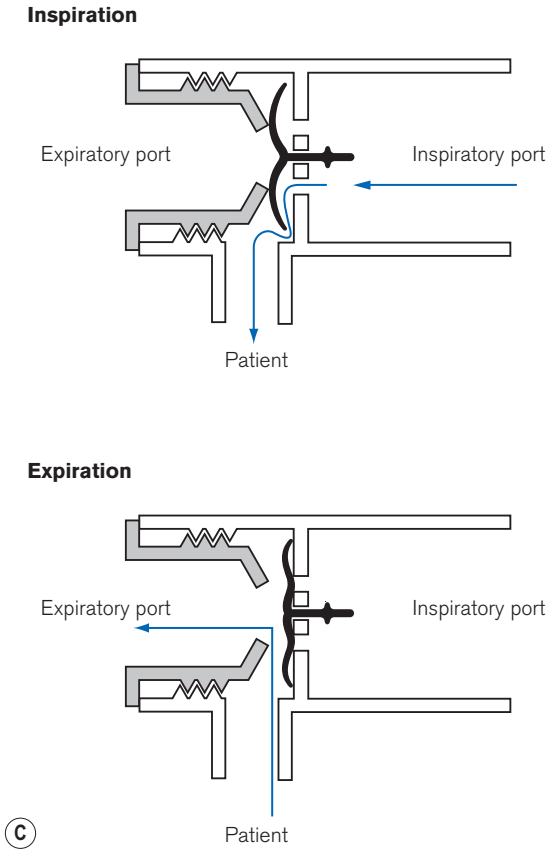


Figure 8.4 **A.** Ambu single-shutter valve. **B.** A SPUR II (single patient use resuscitator) system. **C.** Working principles: upper part, the valve is pushed onto the expiratory port and occludes it so that gas can enter only the patient port; lower part, exhaled gas pushes the valve against the inspiratory port, occluding it so that the gas can escape only through the expiratory port.

The guide stem and flexible shutter are clearly visible through the transparent valve body and their 'to and fro' movement is an indicator of correct function.

The self-inflating bag supplied with this valve is made from silicone. The single use version has been upgraded from the original Spur system by the addition of extra ribbing to increase its recoil so as to allow a faster rate to be applied. The gas inlet is fitted with the two pressure relief valves as described before.

Laerdal pattern valve

This high-efficiency non-rebreathing valve (Fig. 8.5) is made in three sizes (adult, child, infant). The valve itself has three components, a duck-billed inspiratory/expiratory valve, a valve body housing inspiratory and expiratory ports and a non-return flap valve sited in the expiratory port. Originally designed by Laerdal, it is now used by many manufacturers for their resuscitators. It may be made for single use only, in which case the valve housing is sealed and cannot be opened. The reusable version made of autoclavable materials may be dismantled for cleaning and sterilizing. However, care must be taken to reassemble all the components correctly as there have been reports of misassembly:

- *Inspiratory phase.* The central duck-billed portion of the main valve opens when the attached self-inflating bag is squeezed, or when a patient inhales through it. Almost simultaneously the outer disc-shaped portion of the valve is pushed against the apertures in the valve body, thus sealing the expiratory pathway.
- *Expiratory phase.* Positive expiratory pressure from the elastic recoil of the patient's lungs causes the duckbilled section of the valve to close, thus preventing rebreathing it into the bag. Escaping gas also lifts the flaps on a non-return valve in the

FURTHER READING

Barnes TA, McGarry WP III. Evaluation of ten disposable manual resuscitators. *Respir Care* 1990;**35**:960–8.

Baskett P, Zorab J. The resuscitation greats. Henning Ruben MD, FFARCS(I), FFARCS. The Ruben valve and the AMBU bag. *Resuscitation* 2003;**56**:123–7.

Corley M, Ledwidge MK, Glass C, Grap MJ. The myth of 100% oxygen delivery through manual resuscitation bags. *J Emerg Nurs* 1993;**19**:45–9.

Hermansen MC, Prior MM. Oxygen concentrations from self-inflating resuscitation bags. *Am J Perinatol* 1993;**10**:79–80.

Hess D, Simmons M. An evaluation of the resistance to flow through the patient valves of twelve adult manual resuscitators. *Respir Care* 1992;**37**:432–8.

Hess D, Spahr C. An evaluation of volumes delivered by selected adult disposable resuscitators: the effects of hand size, number of hands used, and use of disposable medical gloves. *Respir Care* 1990;**35**:800–5.

Ho AM, Shragge BW, Tittley JG, Fedoryshyn JN, Puksa S. Exhalation obstruction due to Laerdal valve misassembly. *Crit Care Med* 1996;**24**:362–4.

Kathy L, Johnston RRT, Khalid Aziz MA. The self-inflating resuscitation bag delivers high oxygen concentrations when used without a reservoir: implications for neonatal resuscitation. *Respir Care* 2009;**54**:1665–70.

Quintana S, Martinez Perez J, Alvarez M, Vila JS, Jara F, Nava JM. Maximum FIO2 in minimum time depending on the kind of resuscitation bag and oxygen flow. *Intensive Care Med* 2004;**30**: 155–8.

Smith G. Problems with mis-assembly of adult manual resuscitators. *Resuscitation* 2002;**53**:109–11.

Smith G. Problems with mis-assembly of adult manual resuscitators. *Resuscitation* 2002;**55**:347–8.

Chapter 9

Automatic ventilators

Andrew J Davey

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In order to inflate a patient's lungs adequately with a mechanical ventilator, sufficient pressure must be generated in the respirable gas within a ventilator or resuscitator (positive pressure ventilation) to overcome the elastic recoil of the lungs and chest wall (their elastance) and the resistance to flow within the airways. These may be normal in healthy patients, requiring the generation of only modest pressures for inflation, or may be grossly abnormal in disease, requiring the generation of much higher pressures in order to provide the same degree of ventilation. Furthermore, some surgical procedures may make it more difficult to inflate the lungs, for example, by restricting the movement of the diaphragm, due to posture or internal intervention. An additional factor during anaesthesia is the resistance of the artificial part of the airway, which may increase accidentally, for example, by mucus accumulation or kinking of the endotracheal tube.

A patient's lungs may also be inflated by using negative pressure. The patient's body (from the neck downwards) or thorax only, is encased in a gas tight container to which an intermittent sub-atmospheric pressure is applied. The thorax is 'sucked outwards' causing air/respirable gas to enter the lungs (negative pressure ventilation). Exhalation is achieved passively as a result of the elastance of the lungs and thoracic wall. Purely negative pressure ventilators are not considered in this text; intensive care oscillators are discussed in the succeeding chapter.

POSITIVE PRESSURE VENTILATORS

The last 10 years has seen a radical change in the design of positive pressure ventilators. In the developed world; electronics, microprocessors and miniaturized proportional flow valves replaced some older technology such as mechanical, pneumatic and fluidic controls. The manner in which positive pressure ventilation is employed in anaesthesia has also undergone major changes. Ventilators that obtained the patient's minute ventilation from an anaesthetic machine, delivered it to the patient and then vented it to atmosphere (minute volume dividers) have been superseded mainly by devices that utilize circle systems and low flows. As a result, the plethora of ventilator designs that catered for specific situations have been relegated to history, to be replaced by newer and fewer models that by virtue of their electronic adaptability, can outperform their predecessors. However, there are some basic principles that remain the same.

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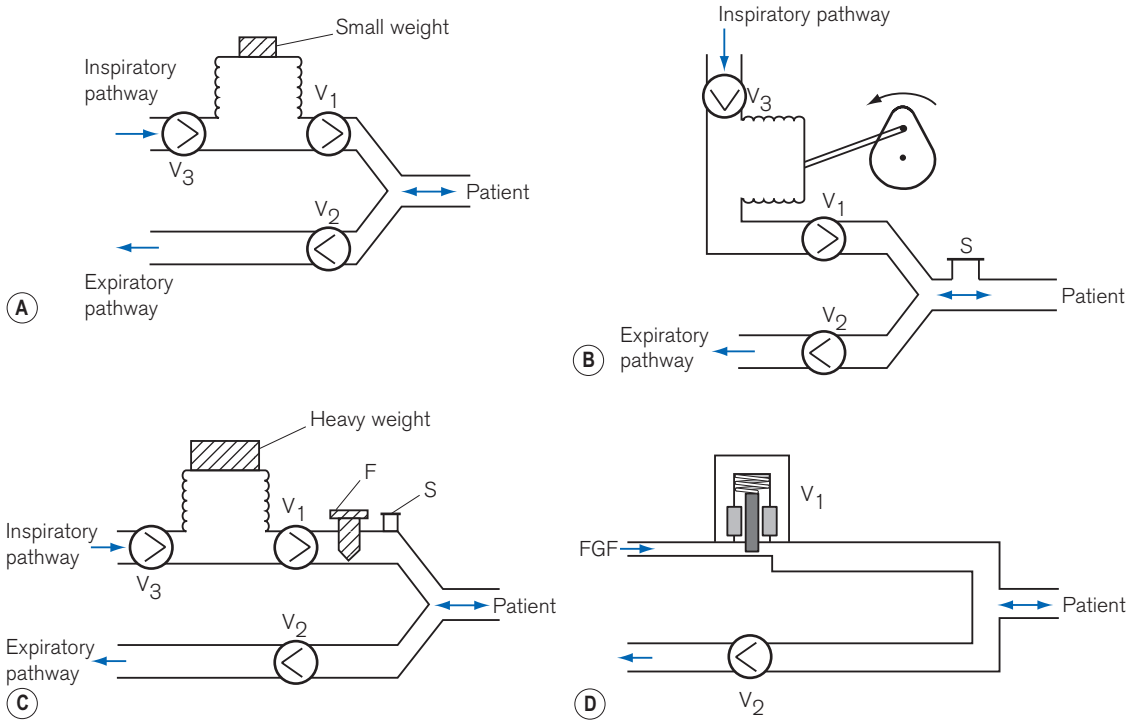


Figure 9.1 **A.** Low-powered ventilator with bellows compressed by a light weight. **B.** High-powered ventilator with bellows compressed by a mechanical arm attached to an electric motor **C.** High-powered ventilator with bellows compressed by a heavy weight **D.** High-powered ventilator using modified high-pressure gas from a pipeline. Components of these systems include V_1, inspiratory valve; V_2, expiratory valve; V_3, non-return valve; F, flow restrictor; S, overpressure relief valve.

Methods of pressure generation

Respirable gas with sufficient pressure to ventilate a patient's lungs may be supplied by a ventilator that obtains gas at atmospheric pressure and compresses and stores this in a bellows or bag by mechanical means (Fig. 9.1A, B and C). Alternatively, a ventilator can be designed to utilize a compressed gas from a cylinder or pipeline supply (Fig. 9.1D) and reduce this to suitable respirable pressures without the requirement of a storage bellows.

Classification of ventilators

A number of attempts have been made to classify ventilators according to their power, efficiency and modes of cycling between inspiration and expiration.

Power

Low-powered ventilators

Low-powered ventilators generate only the modest gas pressures required to deliver reasonable tidal volumes to lungs with normal and near-normal compliances and resistances. These pressures may be insufficient to overcome the increase in airways resistance and/or the reduction in lung compliance that are seen in diseased lungs. As a result of this, the tidal volume delivered may well be less than the volume anticipated. When these ventilators are used, the need to monitor adequacy of lung ventilation must be emphasized. Either expired minute volume or capnography can be used to check that ventilation remains satisfactory throughout a procedure.

With the advent of modern electronics, many ventilators that fit this classification exclusively have become obsolete in the developed world. Some older ventilators and some of those designed for the developing world are so constructed that they can only deliver modest pressures (by using weak springs or light weights to compress the respirable gas in a bellows (Fig. 9.1A)). Such machines are simple to operate, reduce the potential incidence of barotrauma to lungs and, more pertinently, do not require an electrical power supply and are essentially user serviceable.

However, many current electronic high-powered ventilators have a pressure-controlled mode that allows operative characteristics similar to low-powered ventilators. When

used in this mode and the inspiratory pressure is limited to 15–25 cm H_2O via the machine's electronics, the ventilator may be considered as low powered.

High-powered ventilators

In order to prevent a reduction in ventilator performance in the presence of deteriorating lung conditions, a ventilator needs to be powerful enough to overcome the increases in airways resistance and reduction in compliance with little alteration in desired gas flow. These ventilators require also the addition of certain safety features to protect patients with both normal and abnormal lungs from excessive pressures. For example, an overpressure safety valve is always included in the gas pathway to the patient to release any build-up of potentially dangerous pressures that might damage the lungs. Fig. 9.1B shows an example of a typical high-powered ventilator. The pressure-relief valve (S) can either be pre-set (usually at 4.4 kPa/45 cm H_2O) or, in more sophisticated machines, can be adjustable (up to 7.8 kPa/80 cm H_2O) to cope with severe conditions such as asthma and the adult respiratory distress syndrome. Higher-pressure relief settings, however, equate to increased risk of barotrauma.

Those high-powered ventilators that always generate high pressure of gas in the ventilator system prior to its delivery (by using powerful springs, heavy weights or a pipeline gas source (Fig. 9.1C and D)), require the presence of a further safety device, a flow restrictor (see below), in the inspiratory pathway. This reduces the flow to the patient and prevents too rapid a build-up of pressure in the lung.

Alternative classifications

A popular classification with British anaesthetists has been described by Mushin.[1] Those ventilators that by their design produce a pressure sufficient only to ventilate normal or mildly abnormal lungs are classified as *pressure generators*, i.e. the tidal volume delivered to the patient is limited by the pressures generated. Those ventilators that develop pressures sufficiently high enough to deliver a desired flow even to grossly abnormal lungs are deemed *flow generators*. However, as most other electromechanical devices in common usage are described in terms of power, the author prefers the first classification.

Efficiency of ventilators

This may be defined as the ratio of the intended tidal volume (as determined by the settings on the ventilator) over the actual delivered tidal volume. For example, when a ventilator acts on a bellows containing patient gas at atmospheric pressure, the gas undergoes a degree of compression in order to raise the pressure sufficiently to provide an inspiratory flow. Part of the bellows travel is taken up in compressing the gas. If the bellows travel is calibrated for volume, it becomes apparent that the tidal volume actually delivered is less than that indicated on the bellows scale. The greater the pressure required to ventilate a patient's lungs, the greater will be the amount of gas lost in compression. This type of ventilator is regarded as relatively inefficient, as the discrepancy between anticipated and delivered tidal volumes may be as great as 25% in patients with significant pathological lung conditions.

Furthermore, the effective inspiratory time is shortened as, initially, time is lost in compressing the gas to the required pressure. Inefficient ventilators (which include most anaesthetic ventilators that supply circle systems) may well require validation of the delivered tidal volume, using a spirometer or capnograph. With the advent of more sophisticated measurement of flow and electronic feedback to the ventilator, the compliance of this type of system, and, therefore, the compression volume, can now be calculated and automatic adjustment made for most of the apparent 'lost' volume.

More efficient ventilators utilize respiratory gas already under a pressure greater than that required to ventilate a patient's lungs, so that the gas is already compressed prior to being released and, therefore, none is lost in a 'compression volume'. It is important to grasp this concept, as there may be a marked difference in the anticipated performance of ventilators.

Inspiratory characteristics of ventilators

Ventilators may produce a variety of pressure waveforms and inspiratory flow characteristics depending on the method of generation of respirable gas pressure and the resistance to flow that the gas meets during delivery of the intended tidal volume.

Low-powered ventilators

Low-powered ventilators deliver gas at modest pressure. This pressure is normally constant (Fig. 9.2A) and will produce an inspiratory flow rate of gas that is greatest in early inspiration, when the pressure differential between the ventilator and the lung is wide, but that slows during inflation of the lung as the pressures approximate (Fig. 9.2B).

High-powered ventilators

High-powered ventilators function by delivering a sufficiently high driving gas pressure to overcome most abnormal resistance without significantly altering the flow from the ventilator, which remains largely unaltered from the intended settings.

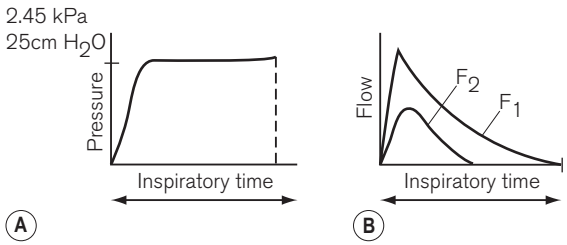


Figure 9.2 Inspiratory characteristics of low-powered ventilators. **A.** Constant pressure (weighted bellows); **B.** Flow pattern in normal (F1) and abnormal (F2) lungs.

Inspiratory characteristics will depend on a number of factors. The high driving pressure from a pipeline source or heavy weighted/spring-loaded storage bellows requires some form of flow restriction to prevent too rapid a rise or an excessive pressure transmitted to the patient's lungs that could produce barotrauma. This may take the form of a *fixed* orifice restrictor. Here the flow will be constant (pipeline supply or weighted bellows) or gradually decreasing (spring loaded bellows) as the tension in the spring reduces with emptying of the bellows (Fig. 9.3A). However, practically in the case of the latter the reduction is insignificant and flow is virtually constant.

In more sophisticated ventilators, the inspiratory flow valve acts as a *variable* flow restrictor (Fig. 9.3B). These are able to respond to user-programmed inspiratory flow patterns.

Ventilators may be designed to force their bellows to be compressed either mechanically, via a linkage from a suitable power source, or pneumatically, by placing the bellows in a gas-tight container into which a pressurized gas source is fed (bag-in-bottle arrangement). The bellows in this type of ventilator normally fills with gas at near atmospheric pressures, so that when it is compressed, the pressure developed rises as it overcomes the resistive properties of the lungs. The resultant pressure and flow waveforms are dependent on the type of mechanical linkage (e.g. rotating cam/linear motor) or the type of pneumatic drive producing any of the waveforms seen in Fig. 9.3. Although these ventilators are classified as high powered, they do not require flow restrictors as there is no initial very high-pressure source present. However, they do need overpressure relief valves to protect against high pressures that might develop unexpectedly.

In either type the delivery of the intended tidal volume is assured owing to the power developed by the ventilator (unless the pressure relief valve opens). More sophisticated ventilators will provide an alarm signal if this occurs.

Great store has been placed on the ability of different flow waveforms to increase ventilatory efficiency in various clinical situations. However, in anaesthetic practice the claimed advantages are less demonstrable.

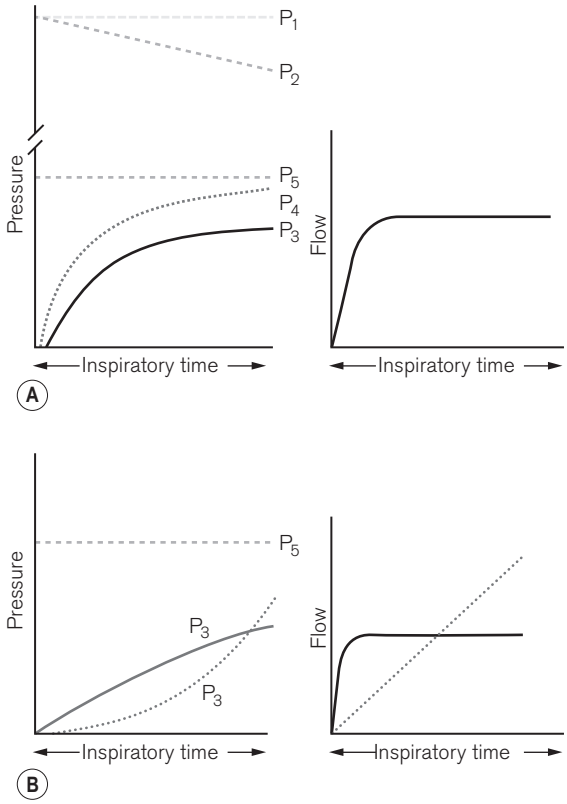


Figure 9.3 Inspiratory characteristics of high-powered ventilators. **A.** Constant high-pressure generation (P_1) well in excess of that required to ventilate abnormal lungs (heavy weight or pipeline gas supply) with *fixed-performance flow restrictor*. High-pressure generation produced by a bellows compressed by powerful springs provides a gradually declining pressure as the bellows empties (P_2 above). However, this is insufficient to affect the performance of the ventilator, which develops pressures and flows as if it were constant high-pressure generation. Flow patterns generated by P_1 and P_2 are hence similar in a given lung scenario. **B.** Constant high-pressure generation using a *varying orifice flow-restrictor* that is electronically controlled to provide different inspiratory flow patterns from the same ventilator (solid line = constant flow; dotted line = increasing flow.).
P_3, pressure rise downstream of restrictor as a result of flow to normal lungs; P_4 as above but to abnormal lungs; P_5, inspiratory safety valve release pressure.

Classification of ventilators according to cycling

Intermittent automatic ventilation of the lungs consists of two phases: inspiratory and expiratory. A ventilator is said to *cycle* between the two phases.

Inspiratory cycling

During the inspiratory phase, the ventilator delivers (a) a *volume* of gas into a patient's lungs, which takes place over (b) a given period of *time*, producing (c) an increase in airways *pressure*. There may also be a change in the pattern of (d) *flow* (inspiratory waveform) at some stage in inspiration. However, the ventilator can allow only one of these variables (a–d) to terminate the inspiratory phase when its predetermined value is reached. As all four variables are present in every inspiratory phase, it is sometimes difficult to decide which one is the principal determinant of inspiratory cycling.

Volume cycling

A ventilator designed to use this method of inspiratory cycling recognizes the point at which a pre-determined volume of gas has left the ventilator and switches its internal mechanism to allow exhalation to occur. Some volume-cycled ventilators have a variable performance restrictor which slows down the inspiratory gas flow, introducing some element of timing during inspiration. However, volume remains the primary determinant of inspiratory cycling.

Time cycling

Mechanical, pneumatic or electronic timers may be used to control the operation of the inspiratory and expiratory valves that govern the cycling of the ventilator which can, therefore, function independently of the delivered tidal volume. This increases the sophistication of the ventilator. For example, not only can widely differing volumes be delivered in a given time frame, but it may also allow a tidal volume to be delivered early in the inspiratory cycle, followed by a pause to allow better distribution of the gas prior to the start of the expiratory phase. Time cycling is now incorporated in most new ventilators.

Pressure cycling

Pressure-cycled ventilators sense a predetermined airway pressure in order to terminate the inspiratory phase. However, if the airway resistance increases and/or if compliance of a patient deteriorates a pressure-cycled ventilator will deliver a reduced tidal volume at the pre-set cycling pressure. The performance of these ventilators is thus very variable.

Flow cycling

Recognition of flow pattern changes has been used to cycle ventilators. However, this method is rarely employed nowadays.

Expiratory cycling

The expiratory phase may be similarly terminated by one of the above-mentioned variables. For example:

- an expiratory *volume-cycled* ventilator may have a mechanism for terminating the expiratory phase when the reservoir bellows has filled to the desired tidal volume required for the next inspiration
- an expiratory *pressure-cycled* ventilator would be able to identify a selected airways pressure at the end of exhalation that would trigger the next inspiratory phase
- an expiratory *flow-cycled* ventilator would switch to the inspiratory phase when the desired flow rate at the end of exhalation was reached, or
- an expiratory *time-cycled* ventilator that terminates the expiratory phase after a predetermined time. This is the most versatile type as its phase may extend beyond the end of patient exhalation, unlike the others. It is, therefore, the most popular method of expiratory cycling and is achieved by using electronic or pneumatic timers within the ventilator to switch phases.

Further explanations are included in the individual ventilators mentioned below.

Ventilators may use one of the methods described above for inspiratory cycling and another for expiratory cycling, depending on the method of construction, and in some of them, limits may be set to one or more of the above functions.

Cycling mechanisms in ventilators

Gas flow to and from the patient from a ventilator (cycling) is usually controlled by a series of one-way valves that are operated and synchronized either:

- mechanically
- electronically, or
- pneumatically.

Examples of these will be described where appropriate in the section on individual ventilators.

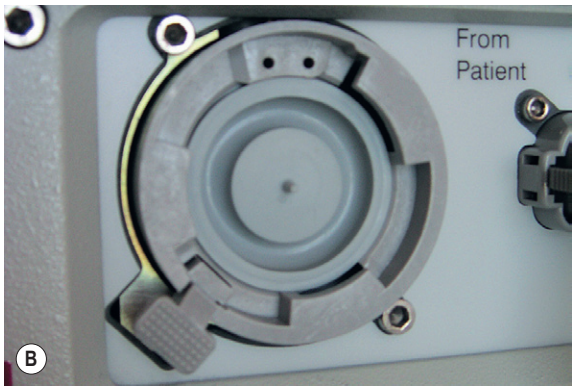
Ventilation modes

The terminology used to describe the way in which a ventilator combines its power capability and cycling to deliver a tidal volume has previously been almost self-explanatory despite manufacturers coining their own names.

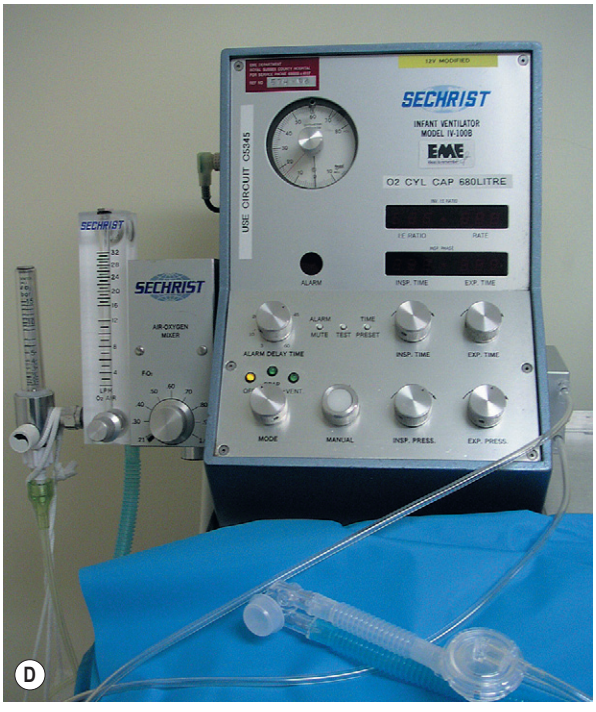
Originally, ventilators were used to ventilate apnoeic/paralyzed patients. Where a desired tidal volume and rate was delivered by a high-powered ventilator, this was usually referred to as *controlled minute ventilation* (CMV), *volume-controlled ventilation* (VCV) or *volume ventilation* (VV).

Low, or high-powered ventilators that have a pressure limit for the delivery of a tidal volume, were said to deliver *pressure-controlled ventilation* (PCV), *pressure ventilation* or *pressure mode*.









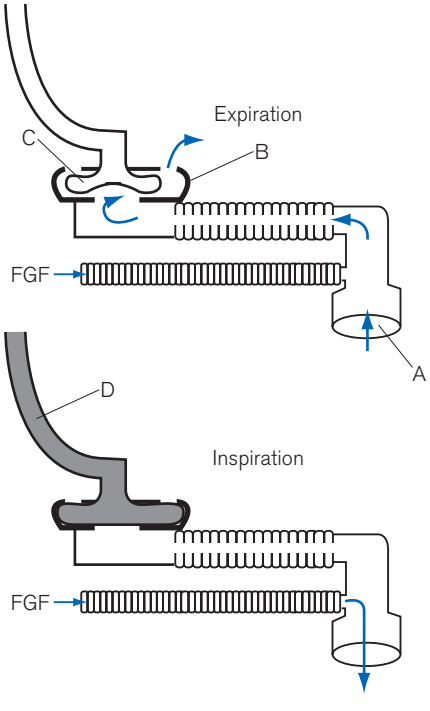


Figure 9.5 **A.** Bird VIP neonatal ventilator; **B.** Silicone diaphragm of Bird VIP acting as a variable flow valve; **C.** Bird VIP Piston from flow valve that moves the diaphragm. **D.** Seechrist infant ventilator; **E.** schematic diagram of the 'mechanical thumb'. A, patient connection; B, expiratory valve; C, deflated pneumatic valve; D, gas supply tube to pneumatic valve, which is now inflated.

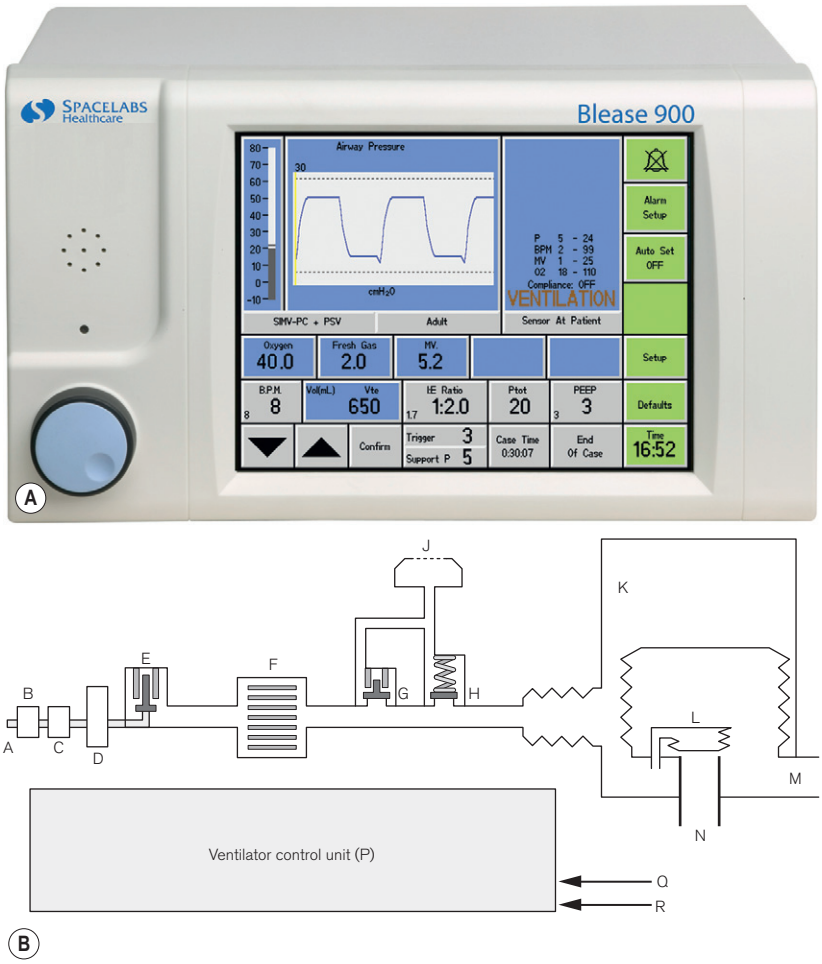


Figure 9.9 **A.** Blease 900. **B.** Working principles of Blease 900. A, drive gas input; B, gas filter and water trap; C, on/off switch; D, pressure regulator; E electronically operated proportional inspiratory flow valve; F, pneumotachograph; G, electronic expiratory/multifunction valve; H, mechanical overpressure relief valve; J, driving gas exhaust; K bellows assembly; L, pneumatic valve; M, pathway to breathing system; N, patient gas exhalation port; P ventilator control unit; Q, fresh gas flow sensor; R, patient gas pressure and flow sensor.

these with pressure-supported spontaneous breathing, e.g. SIMV-PC+PSV. An inspiratory pause can be dialled in, and there is the return of the 'sigh' option. The latter in default mode delivers a breath larger than tidal volume by 10% every 10th breath, although both values may be altered by the user. The ventilator may be used to ventilate paediatric patients using the appropriate flow sensor and ventilator settings.

Ventilator controls The front of the ventilator casing houses a flat panel TFT touch screen, similar to that found on a modern computer (Fig. 9.9A). Also, there is a rotary knob (Trak wheel) on the bottom left of this. Adjustable parameters may be selected by touching the appropriate box and using the up/down arrows to select the desired value. Alternatively the Trak wheel may be rotated to select a parameter. When pushed inwards (clicking) it highlights that value which may then be changed by further rotation. This value is confirmed by again 'clicking' on the wheel.

Many other features, including back-up ventilation in case of apnoea, and safety systems are available. These are not described here, but can be found in the manufacturer's user manual.

The ventilator design can, therefore, be classified as a high-powered, time-cycled 'bag squeezer' ventilator.

GE Healthcare 7900 Smartvent

This is another example of a pneumatically driven microprocessor controlled bag squeezer (Fig. 9.10). The operating principles are similar (although with minor variations) to that described above in that high-pressure driving gas is passed through an electronically controlled proportional

flow valve to externally manipulate a bellows arrangement that contains patient gas. In this device the same proportional valve is used to deliver a bias flow to produce PEEP during the expiratory phase by acting on the passive expiratory valve (Fig. 9.10 B). The electronic control is handled by a programmable microprocessor to deliver a wide selection of respiratory modes. As with most anaesthetic ventilators of this type, these modes are similar to those found on a typical ITU device. Hence, there is VCV and PCV, both with SIMV facility and the ability to allow and/or pressure support a spontaneous breath. The GE version of pressure support ventilation (PSV Pro) may also be used to augment intended spontaneous respiration. This mode also has a back-up for unexpected apnoea. If the patient does not take a breath within the pre-set apnoea delay time, the apnoea alarm will activate and the ventilator will automatically switch to the back-up mode, which is SIMV-PC.

With the advent of fast response proportional valves, all these modes may be used in paediatric as well as adult patients. Different manufacturers often add subtle changes to various modes and also use slightly differing nomenclature for similar modes to that of their competitors. For example, the 7900 has a 'volume guarantee' mode in PCV. Here, the clinician sets an intended tidal volume, half of which is delivered in the first breath using volume controlled mode. The pressure generated by this breath is then used to calculate the level of pressure control for subsequent breaths which are gradually stepped up with the aim of achieving the set tidal volume within seven breaths. Hence all breaths after the first are delivered using

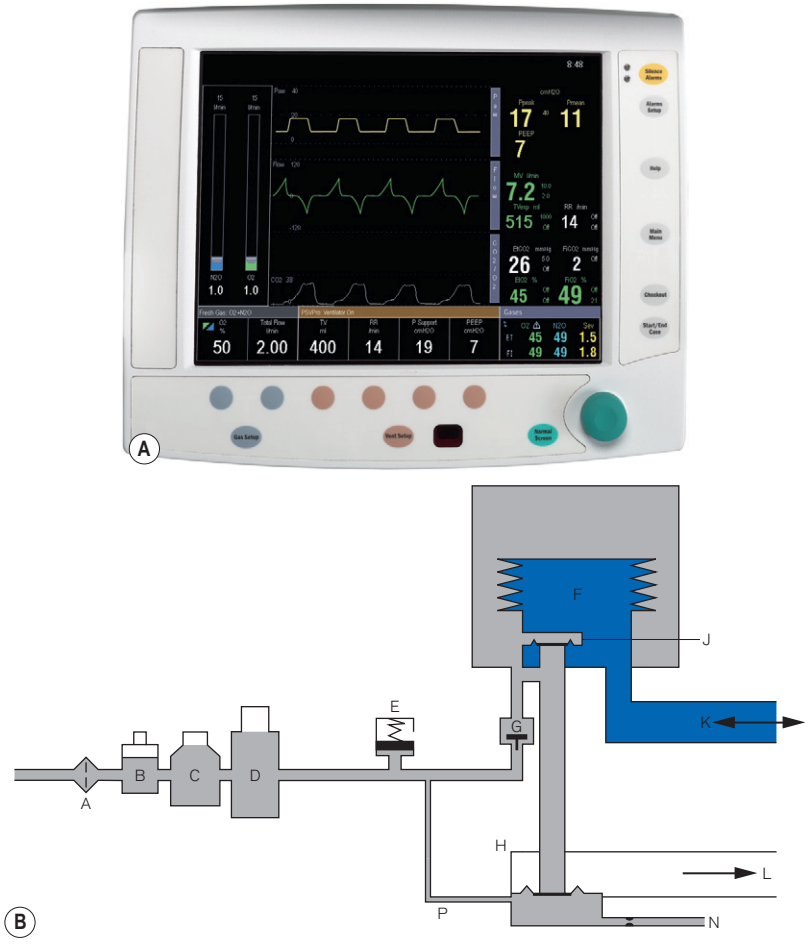


Figure 9.10 GE Smartvent 7900 ventilator.
A, ventilator control panel as part of the anaesthetic machine user interface (Aisys, GE Healthcare, shown here with US gas colouring). Image courtesy of GE Healthcare. **B,** working principles: (A) drive gas input filter; (B) Gas Inlet Valve – solenoid, opens when ventilator is on, closes under fault conditions such as system overpressure; (C) drive gas pressure regulator (output at 170 kPa); (D) flow control (proportional) valve; (E) mechanical overpressure safety valve (110 cm H_2O); (F) bellows; (G) drive gas check valve (3.5 cm H_2O bias); (H) exhaust valve manifold; (J) bellows pressure relief 'pop off valve'; (K) connection to circle system; (L) patient and drive gas exhaust to scavenging; (N) control bleed to atmosphere; (P) piloting pressure to exhaust valve.

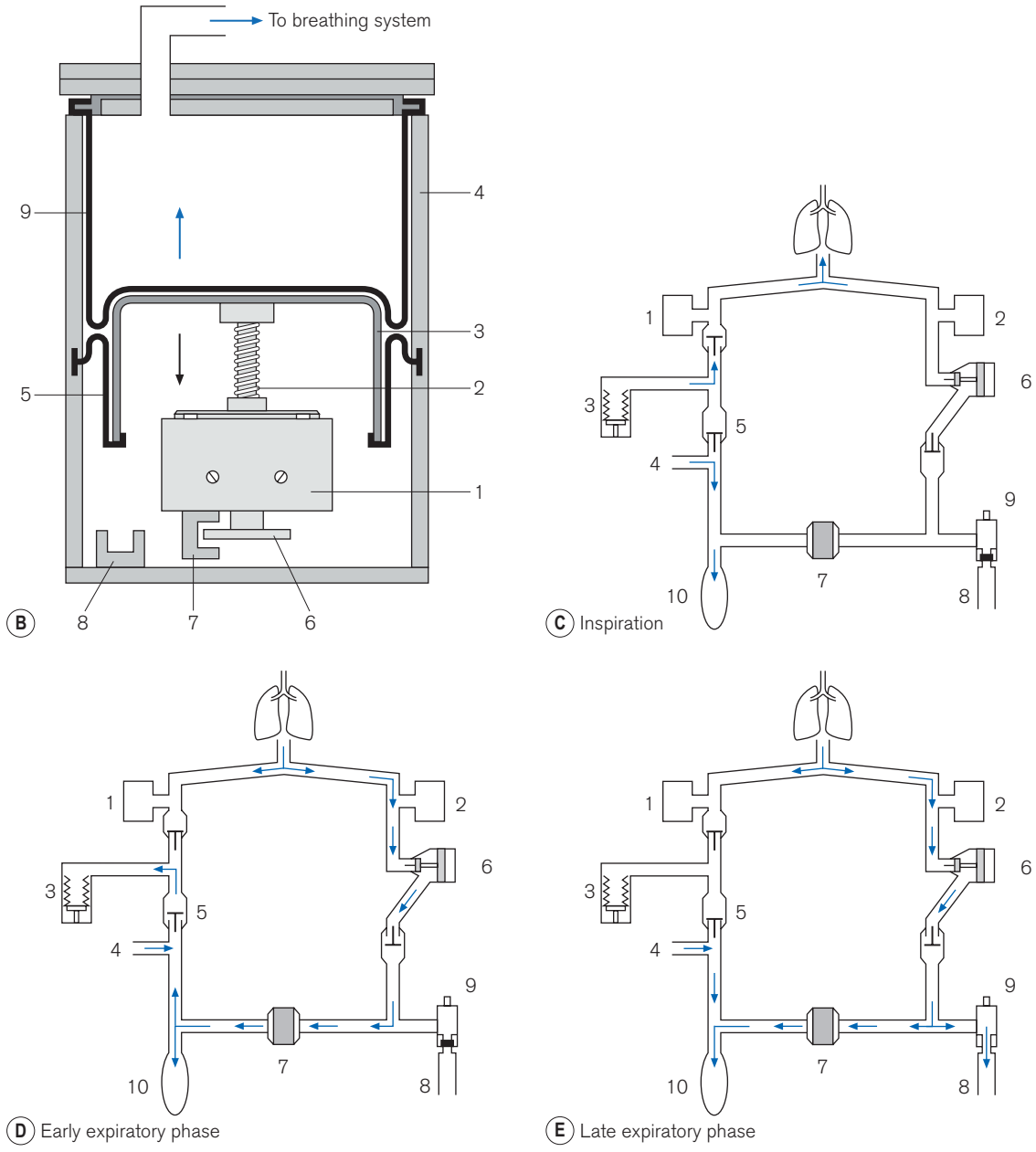


Figure 9.11, cont'd **B.** Working principles: (1) electric motor; (2) rod with screw thread; (3) piston; (4) cylinder; (5) rolling rubber seal; (6) incremental encoder; (7) sensor; (8) light barrier; (9) ventilator bellows. **C.** inspiratory phase: (1) inspiratory flow transducer; (2) expiratory flow transducer; (3) ventilator; (4) fresh gas; (5) fresh gas decoupling; (6) valve controlling P_{max}/PEEP; (7) absorber; (8) scavenging; (9) APL valve; (10) reservoir bag. **D.** early expiratory phase. **E.** late expiratory phase. (Fig. 9.11A reproduced with permission from Dräger Medical UK.)

module in the model used. The basic model, the Fabius CE allows volume-controlled ventilation only. The top of the range Primus has the facility for both VCV and PCV, either as stand-alone features or with the provision for spontaneous breathing with and without pressure support in these modes. It also allows a purely spontaneous respiration mode with triggered pressure support. The control unit shown is a mid-range model, the Fabius GS. Like the ventilator described above, it has a rotary control (bottom right) to alter selected variables. On the left of the unit are the keys that select the mode of ventilation. The middle of the unit has a thin film transistor (TFT) screen that displays all the relevant information. To the right of this are two banks of keys to select the alarms, menu set-up, home, alarm silence and standby. The home key restores the screen to the default after any submenu called up is no longer required and the standby key stops the ventilator and keeps any ventilatory parameters selected for use again.

Ventilator module (Fig. 9.11B) Unlike the ventilators described above, this device does not require a pressurized source of gas as its power source. The ventilator has an electric motor (1) with a hollow spindle. The inside of the spindle has a screw thread. A rod (2) with a matching thread passes through the spindle. When the electric motor spins, the spindle rotates and the action of the two threads, which are interlocked, causes the rod to move through the spindle. This movement is referred to as either a recirculating ball screw or a worm drive. One end of the rod is connected to a piston (3) that moves backwards and forwards inside a cylinder (4), depending upon the direction and duration of current flow in the electric motor. The head of the piston is fitted with a rolling neoprene seal (5) so that on the downstroke it is capable of producing a sub-atmospheric pressure to the bellows that sits above it. The position of the piston rod at any one time is sensed by a high-resolution incremental encoder (6) and allows precise volume (0.03 ml) delivery. The encoder consists of a metal disc that has 1024 perforations around the edge. When the electric motor is working, this disc spins between two arms of a sensor (7) that counts the passage of the perforations and then calculates the linear movement of the piston rod. At the bottom of the cylinder there is a light barrier to detect the lower stop position of the piston.

Interesting features

- *Inspiration* (Fig. 9.11C). During the inspiratory phase the ventilator delivers the intended amount of volume to the patient. It does this by diverting the FGF from the anaesthetic machine via a decoupling valve (5) into the reservoir bag and not the patient. Furthermore, the delivered tidal volume from the ventilator enters the breathing system downstream of the absorber (which is isolated) and, therefore, minimizes the compression volume of the inspiratory pathway. The reservoir bag will be seen to expand with the addition of fresh gas, which may be a little unnerving for those not familiar with this system. There is a flow transducer in the inspiratory pathway that measures gas flow, which is then displayed on the control unit.
- *Exhalation* (Fig. 9.11D). The expiratory travel of the piston is fixed by the I/E ratio of the ventilator. The sub-atmospheric pressure created by the downstroke of the ventilator sucks in gas from both the reservoir bag and the exhalation volume from the patient. Towards the end of the exhalation phase, the reservoir bag fills and surplus gas is dumped through the scavenging port. Again the reservoir bag will be seen to move. In the top of the range machine, information from the expiratory flow transducer is passed to the microprocessor, which in turn causes the movement of the piston backstroke in the ventilator to match the expiratory flow.

Paediatric mode When these ventilators are switched on, the control unit software performs a leak and compliance test (except for the base model workstation). Compliance compensation, along with the low compliance of the bellows and breathing system, allows accurate delivery of small tidal volumes if required. More commonly, the ventilators may be used in the pressure support made that compensates for any small leak caused by an uncuffed endotracheal tube.

The ventilator design can, therefore, be classified as a high-powered, high-efficiency, time-cycled 'bag squeezer' ventilator.

Intermittent blowers

These ventilators are driven by a pressurized source of gasses or air, at a pressure of 250–400 kPa (37.5–60 psi). The driving gas pathway is very small with a low internal compliance making this type of device very efficient. The major component is an electronically timed and activated proportional flow valve or a pneumatically timed oscillator that divides the driving gas into tidal volumes the size and rate of which can be adjusted. Sophisticated ventilators such as those used in intensive care and anaesthetic workstations make use of a proportional flow valve (see above). Automatic resuscitators and more basic anaesthetic ventilators use the pneumatic oscillator principle (see below) as this is cheaper, does not require the same sophistication of operation and is powered by the driving gas requiring no electrical supply.

Pneumatic oscillator A typical example is seen in Fig. 9.12. The diagram is a very simplified version and does not attempt to show the detailed pneumatics that are essential for its function. Driving gas enters at point (1). It divides into three pathways. The main one passes to a cylinder that contains a shuttle (2), which travels between the ends of the cylinder. In the inspiratory phase (Fig. 9.12A), the driving gas passes into the cylinder and

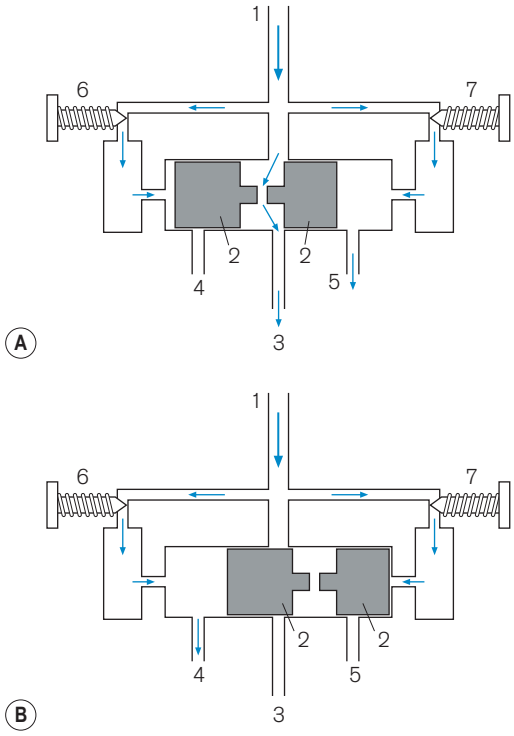


Figure 9.12 A pneumatic oscillator. **A.** Inspiratory phase. **B.** Expiratory phase. (1) Driving gas; (2) shuttle; (3) gas pathway to patient; (4) inspiratory timer vent; (5) expiratory timer vent; (6) inspiratory timer; (7) expiratory timer.

through a hole in the shuttle into the gas pathway (3) to the patient. The other two pathways supply two pneumatic timers (6) (inspiratory) and (7) (expiratory), each of which has a needle valve that regulates flow to the timer mechanism at the relevant end of the cylinder. When the flow causes sufficient build up of pressure in the inspiratory timer, the shuttle is forced to the opposite end of the cylinder (Fig. 9.12B) and in doing so causes three events:

1. It blocks off the flow of driving gas through the cylinder terminating the inspiratory flow.
2. It opens a vent (4) to open on the inspiratory side of the cylinder that allows the pressure in the inspiratory timer to be released.
3. It causes the shuttle to occlude the expiratory timer vent (5) so that a pressure can build up to reverse the direction of the shuttle and terminate the expiratory phase.

Working principles of pneumatically controlled intermittent blowers

A generic line diagram of a typical pneumatically powered and controlled 'intermittent blower' is shown in Fig. 9.13. High-pressure driving gas (300–400 kPa) is connected to the device at point (F). When the main pneumatics on/off switch (G) is turned on, gas flows through it to a regulator (H) that reduces the driving pressure to approximately 275 kPa. Gas flows on to the oscillator (J). The output from this passes through a variable flow restrictor and exits the device to be attached to a breathing system. The delivered tidal volume is a function of the inspiratory timer (K), which is calibrated in seconds, and flow restrictor (R), which is calibrated in l s^{-1}. The respiratory rate is determined by the cycle time: inspiratory time (adjusted at K) plus the expiratory time (adjusted at L).

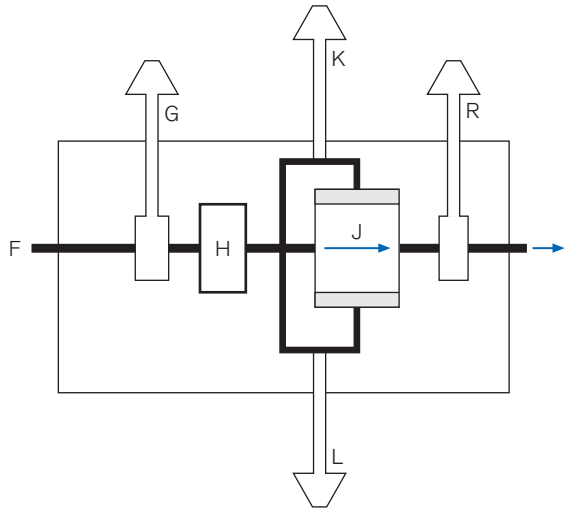


Figure 9.13 Schematic diagram of a pneumatically controlled intermittent blower (see text for details). F, high-pressure driving gas input (300–600 kPa); G, pneumatic on/off switch; H, pressure regulator; J, oscillator; K, variable pneumatic inspiratory timer; L, variable pneumatic expiratory timer; R, inspiratory flow restricter.

Classification of intermittent blowers

Intermittent blowers are used in four different ways (Fig. 9.14).

Basic resuscitators

The simplest design is used as a basic resuscitator (Fig. 9.14A). The working principles are shown in Fig. 9.15A. It has no separate on/off switch and no flow restrictor and a fixed expiratory timer. It has a single control (K) for tidal volume, cycling rate and I/E ratio, which is actually the variable inspiratory timer. Since the flow rate is constant, when the inspiratory time is lengthened, the tidal volume is increased, the cycling rate is reduced and the I/E ratio is prolonged, and vice versa.

An example of this type is the Pneupac adult/child resuscitator (Fig. 9.15B), which, although no longer in production (since 2004), is still widely used.

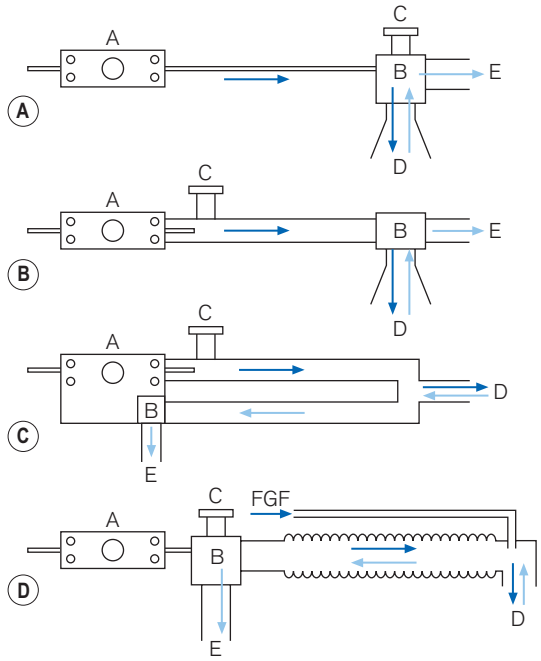


Figure 9.14 Classification of intermittent blowers. **A.** Basic resuscitator. **B.** Sophisticated resuscitator. **C.** Ventilator for intensive care. **D.** Anaesthetic ventilator for Mapleson D system. A, resuscitator/ventilator; B, patient valve; C, overpressure relief valve; D, patient pathway; E, expiratory pathway.

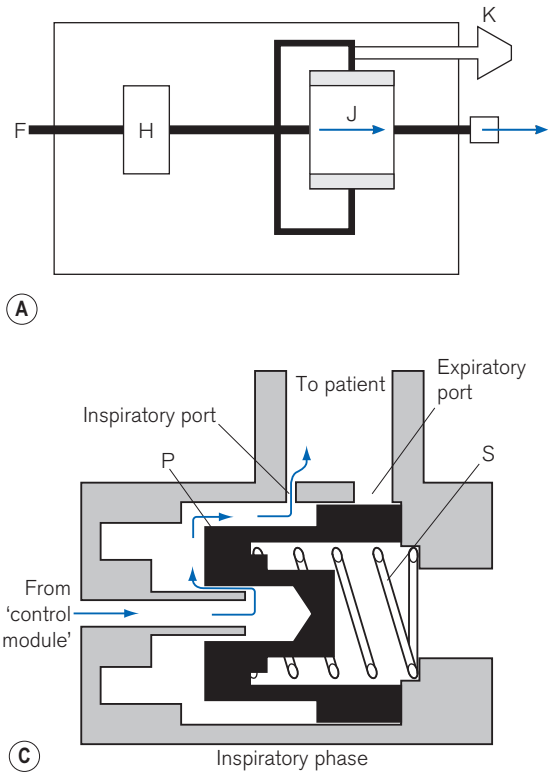




Figure 9.15 **A.** Working principle of a basic resuscitator: F, driving gas; H, pressure regulator; J, oscillator; K, tidal volume and rate control. **B.** A basic resuscitator: the Pneupac adult/child resuscitator. The overpressure relief valve with its red cap is seen connected to the patient valve. **C.** Working principles of the Pneupac patient valve (inspiratory phase): P, piston; S, spring.

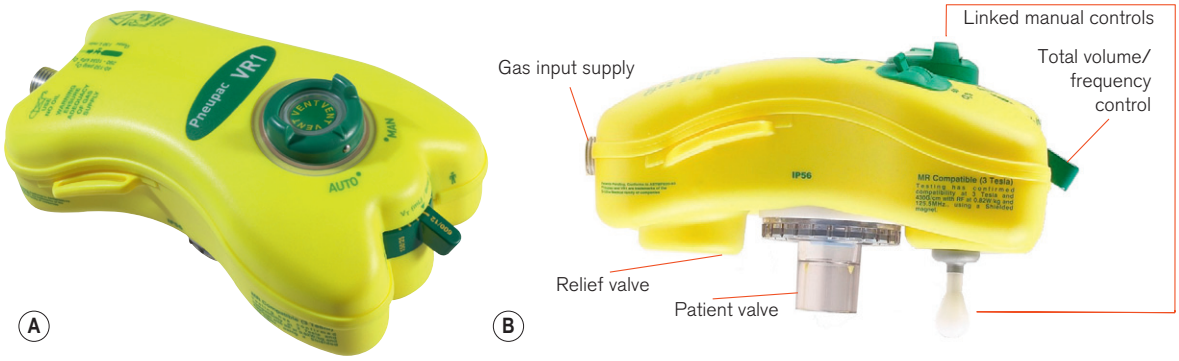


Figure 9.16 (A). The Pneupac VR1 top view. (B). Side view showing the patient valve for connection to a facemask / endotracheal tube.
Photograph courtesy of Smiths Medical, UK.

The working principles of the patient valve are explained in Fig. 9.15C.

Sophisticated resuscitators

This basic model has been superseded by the Pneupac VR1 (Fig. 9.16), which now has an on/off switch, *manual* mode to comply with International Liaison Committee on Resuscitation (ILCOR) CPR guidelines, a *demand* function to allow spontaneous breathing through the device and *variable flows* across the frequency range to deliver gentler breaths. These devices can also be used in toxic environments (for example hazardous area response teams: HART). The working principle is very much as in Fig. 9.17A (see below), but with controls K, L and R combined in one single control for the user. This allows greater speed in the deployment of the device. The variable flows are also available when pushing the manual control button. The manual control is now a standard feature on most resuscitators to allow compliance with the changing CPR ventilation to chest compression ratios (currently 2/30). Also most basic resuscitators now allow direct connection of the patient valve to the resuscitator so that it can be used on top of the mask/tracheal tube (Fig. 9.16B). Alternatively, a length of wide bore tubing may be placed between the patient valve and resuscitator (see Fig. 9.17A).

More features may be added to a resuscitator to increase its scope. However, it then starts to resemble an ITU ventilator. An example is the Pneupac Ventipac (Fig. 9.17A). The increased sophistication may be seen in the line diagram Fig. 9.17B. The output from the oscillator (J) is passed via a variable flow restrictor (R) on to two coupled needle valves (P) and (Q), operated by a switch (O). If P (air mix mode) is selected, the driving gas is passed into a Venturi that entrains a fixed amount of ambient air from S (this port has a non-return valve) and the total flow is fed into the patient breathing system.

If Q (no air mix) is selected, both needle valves are activated. The bore of Q is such that its output matches the entrainment from S to supplement the flow from P so that no air is entrained and the delivered content is 100% driving gas (usually oxygen). The output of the ventilator is connected to the wide-bore hose of breathing system, the other end of which is attached to a light-weight low-resistance Laerdal pattern non-rebreathing valve. The device has separate controls for inspiratory time, expiratory time and inspiratory flow rate and so is able to provide greater ventilatory flexibility than the base model.

In addition, should the patient attempt to breathe, a demand detector (M) senses the pressure in the breathing system via a pilot line (W) and triggers the demand valve (N). This takes its gas supply from the high-pressure gas inlet upstream from the pneumatic switch (G) and allows the demand valve to function even if the ventilator is switched off. If the latter is switched on, however, the demand valve operates in conjunction with the oscillator to integrate this signal and to extend the expiratory phase as a function of the spontaneous tidal volume up to a maximum time dictated by the frequency settings of the ventilator. Thus, if the patient demands a high flow for a short duration or low flow from a longer duration (i.e. similar tidal volumes), an equal expiratory time will be allowed before the next breath. The cumulative effect of successive spontaneous breaths by the patient causes the ventilator to become inhibited, although, in fact, this is only on a breath-by-breath basis. Inhibition starts at 150 ml and increases to a full inhibition of 450 ml. The level of spontaneous breathing required to fully inhibit the oscillator is fixed at that of the typical adult breathing spontaneously. This is taken as tidal volume of about 450 ml at 12 to 16 breaths per minute. Higher spontaneous ventilation rates can readily be taken and will result in complete inhibition of the ventilator. Lower rates will only give partial inhibition, but providing the demand flow is above 15 l min^{-1} the ventilator will still interact with the patient and synchronize its ventilation pattern with the spontaneous breathing. The ventilator has a variable pressure relief valve (T). It also has a battery operated multi-functional pressure alarm and pressure gauge.

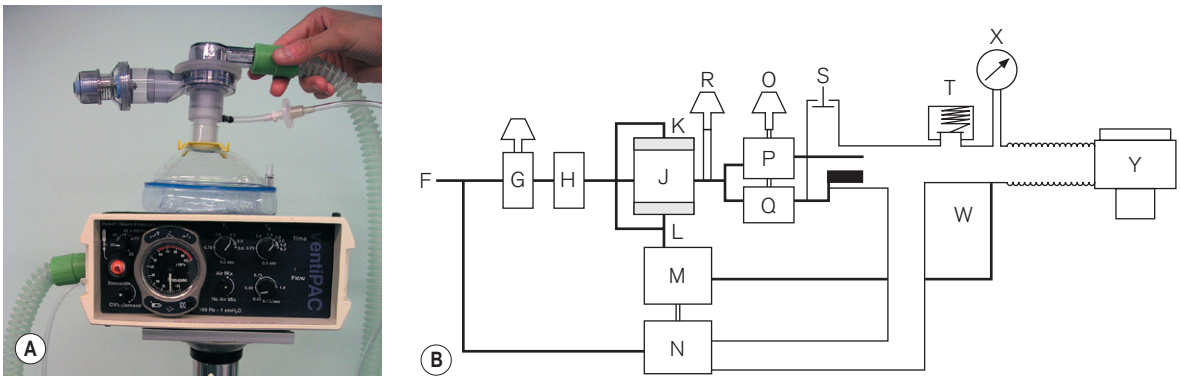


Figure 9.17 **A.** A Pneupac Ventipac ventilator with the airways pressure line, Laerdal pattern non-rebreathing valve with a PEEP attachment. **B.** Working principles: F, driving gas input; G, on/off valve; H, pressure regulator; J, oscillator; K, inspiratory timer; L, expiratory timer; M, demand detector; N, demand valve and demand gas pathway; O, air mix selector switch; P, needle valve for airmix; Q, needle valve to supplement flow from F for no airmix; R, variable flow restrictor; S, air entrainment port with non-return valve; T, variable pressure relief valve; W, pilot line to demand detector; X, airways pressure display; Y, Laerdal pattern non-rebreathing valve.

Intensive care ventilators

Fig. 9.14C shows a very basic line diagram of an intermittent blower used as typical intensive care ventilator. These are discussed in more detail in Chapter 10.

Ventilators for anaesthetic breathing systems

Fig. 9.14D illustrates the use of an intermittent blower with an anaesthetic breathing system. The patient valve may be placed adjacent to the ventilator and the output connected to a breathing system (Mapleson D or circle system) in place of the normal reservoir bag (see Chapter 5). It must be remembered that sufficient length of wide-bore hosing must be used between the ventilator and the breathing system to prevent any driving gas from diluting the anaesthetic intended for the patient. The ventilator that popularized this method in the UK is the Penlon Nuffield 200 series (Fig. 9.18).



Figure 9.18 Penlon Nuffield Series 200.

Jet ventilation

Conventionally, the lungs of a patient are normally ventilated by providing a seal to the upper airway, so that sufficient pressure may build up to provide movement of gas into the lungs. Alternatively, a high-pressure jet of gas may be directed into the airway without the need for a seal. The kinetic energy of the gas molecules is sufficient to overcome the elastic properties of the lungs and to cause them to expand. Furthermore, the speed of the molecules leaving the jet may act as a Venturi and entrain adjacent gas so as to increase the volume provided. The efficiency of the jet depends on a number of factors.

The jet is at its most efficient in terms of gas delivery and ability to entrain when its path is in a straight line. Sharp bends dramatically decrease both of these. The amount of gas delivered is also increased as the driving pressure is raised.

Jet ventilation may be very useful in situations where the airway is so narrowed that only a small gas delivery device may be passed or in situations where conventional airway management devices would impede the view of a surgeon or the conduct of an operation.

There are two ways in which a high-pressure jet of gas may be used to ventilate a patient.

- Gas exchange occurs by normal tidal ventilation, hence.
- Normal tidal volumes are required for carbon dioxide elimination.
- The ventilation can be carried with a manual injector.
- There is sufficient time during exhalation for conventional methods to be used in the measurement of carbon dioxide elimination.
- Larger movements of gas in and out of the laryngeal inlet may cause sufficient movement of the vocal cords to preclude any operative procedure.
- Inflation pressures can be high, if a manual injector is used without meticulous attention to adequate exhalation, and may easily embarrass cardiac output or cause barotrauma.

High-frequency jet ventilation At rates above 60 and up to 300 cycles per minute, a number of advantages are claimed for jet ventilation. Some of these and other salient points are outlined below:

- more efficient alveolar ventilation with much lower tidal volumes
- a substantial reduction in mean airway and alveolar pressures with a reduced potential for barotrauma
- a minimal disturbance of cardiac output, and subsequent renal function
- a reduction in leaks from broncho-pleural fistulae, often with an improvement in gas exchange
- a more comfortable method of providing ventilatory support for patients in the intensive care unit
- for surgical procedures the target site is more stable and only vibrates
- expensive and unfamiliar equipment is needed
- altering ventilatory parameters causes less intuitive changes in physiological variables.

Inhalational anaesthesia

Current jet ventilation systems do not readily lend themselves to providing inhalational anaesthesia. Although nitrous oxide and oxygen mixture can be used with the ventilators (via a high-pressure blender), there are no commercial high-pressure vaporizers available for the addition of inhalational agents.

Entrainment

As the jet may sometimes act as a Venturi, there is a facility to entrain gas supplied from the machine with the same composition as the driving gas in order to guarantee the desired final gas composition.

Humidification

Conventional hot-water or condenser humidifiers are impractical for use with high-frequency ventilation, as they have too high an internal volume. The small delivered tidal volumes with high-frequency ventilation get 'lost' in the humidifier and are not delivered to the patient. However, warmed distilled water or saline can be fed into the jet line by a peristaltic pump. As this emerges from the jet it is atomized to provide a degree of humidification, which can be varied by adjusting the speed of the pump. Humidification in the Monsoon is served by a hot plate arrangement within the ventilator onto which sterile water is fed.

Adjunct to conventional ventilation

Jet ventilation may be superimposed on top of conventional ventilation. A specially adapted Y-piece is required to allow the jet and the airway pressure line to be passed into the lumen of the tracheal tube. Jetting takes place either throughout the respiratory cycle or just during the expiratory phase where it provides an oscillatory form of PEEP.

REFERENCE

1. Mushin WW, Rendell-Baker L, Thompson PW, Mapleson WW. *Automatic ventilation of the lungs*. 3rd ed. Oxford: Blackwell Science; 1980.

FURTHER READING

Spacelabs IBlease 700/900 User Manual, March 2010.

GE Healthcare 7900 Smartvent™, Specification.pdf. 2005.

GE Healthcare Aisys brochure pdf 2006.

Penlon Nuffield 200 Technical description (200 4tecNV200.ppt), 2004.

Glaister C. Dräger Fabius GS Training notes. Dräger UK; February 2003.

Smiths Medical Pneupac®VR1 Emergency and Transport Ventilator (smoths-medical.com/Pneupac).

Sims Pneupac Principles of operation of ventiPAC Pneupac Ltd.2002 Part No504-2101 Issue 1 07/2003.

Acutronic Technical: Service Manual MONSOON III Software 5.0.

Chapter 10

Ventilation in the intensive care unit

Martin Street

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INTRODUCTION

Originally intensive care unit (ICU) ventilators were adapted from standard anaesthetic equipment by the addition of a separate circuit that allowed the patient to take a spontaneous minute volume over and above that delivered by the ventilator. With the widespread use of microprocessors and sophisticated pneumatic controls, modern ICU ventilators allow both mechanical ventilation and patient spontaneous breathing to take place through the same system.

IDEAL REQUIREMENTS

Despite the ideal of one machine being able to cope with all of the clinical conditions encountered in the ICU, the ventilatory strategies are too diverse for the entire, and often conflicting, requirements to be covered by one machine. Safe and easy-to-use machines are important for modern clinical practice. The numerous ventilators available for use in ICUs have diverse mechanical, pneumatic and electrical components governing their working and behaviour characteristics. An ideal ICU ventilator should provide:

- the ability to ventilate all sizes of patient from neonate to obese adults. However, specific machines have been designed to ventilate neonates and most ICU ventilators are designed only to have the ability to ventilate small infants to large adults
- operational versatility with the ability to provide different patterns of ventilation for varied clinical circumstances. The machines should offer the ability to alter such characteristics as inflation pressure, tidal volume, gas flow, respiratory rate and inspiratory to expiratory (I/E) ratio
- a facility for the patient to breathe spontaneously through the ventilator in spontaneously breathing mode without imposing an increased work of breathing
- the ability to augment patient efforts in spontaneous breathing modes to prevent respiratory muscle fatigue
- the ability to increase the pressure in the inspiratory limb of the patient circuit for the application of positive end expiratory pressure (PEEP) and

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continuous positive airway pressure (CPAP) in spontaneous breathing modes
- the ability to deliver a preset volume with flow characteristics independent of changes in a patient's lung resistance and compliance
- the ability to cope with large leaks from the patient circuit without altering performance: essential for non-invasive ventilation via masks or hoods
- the delivery of precise inspired oxygen concentrations varying from 21 to 100% in all ventilatory modes
- the ability to humidify inspired gasses without changing ventilator characteristics
- the ability to add drugs or additional gasses such as helium and to nebulize bronchodilators into the inspiratory limb without altering the ventilator performance or inhaled oxygen concentration
- accurate and reliable monitoring of patient and ventilator respiratory performance along with alarms should these exceed predefined limits
- patient safety features such as high pressure relief valves and gas supply safety features in the event of either electrical, gas supply or control system failure
- easy to use and intuitive operator controls
- the ability to ensure that patients are not exposed to cross infection hazards by using disposable or easily sterilised patient circuits and expiratory valves
- the ability to work independently of mains electrical or pipeline gas supply to facilitate patient transport
- reliable component parts with infrequent routine maintenance schedules.

DIFFERENCES BETWEEN VENTILATORS FOR ANAESTHESIA AND INTENSIVE CARE

Ventilators used in anaesthesia were primarily designed to replace the minute ventilation of a patient with healthy lungs who is paralyzed or has a depressed respiratory drive as a result of anaesthetic agents and opiates.[1] In contrast intensive-care patients are normally encouraged to breathe alongside the mechanical ventilatory support which is gradually reduced during a slow weaning phase. Additionally at the end of anaesthesia, extubation is more precipitous with the rapid recovery of consciousness and return of spontaneous breathing. ICU ventilators, therefore, have to cope with the complexity of both machine- and patient-initiated respiration.

The pulmonary mechanics in terms of airways resistance and total lung compliance in ICU patients is rarely normal. ICU ventilators have, therefore, evolved to support respiration using such techniques as inverse ratio ventilation and high levels of PEEP, which are rarely required during routine anaesthesia. Ventilation for prolonged periods of time may require humidification techniques other than the simple heat and moisture exchanger commonly found in the operative setting. These alternative techniques such as hot water humidifiers are associated with increased accumulation of fluid in the respiratory circuits, which may compromise ventilator function. Large air leaks from the lungs or around airway tubing are more common in the ICU setting, and monitoring facilities on ICU ventilators have to be adapted to function in these unusual circumstances to prevent an excess of false alarms. Nonetheless, the flexibility of modern ICU ventilators is utilizable and sometimes indispensible in the operating theatre and modern anaesthesia ventilators are increasingly equipped with features of an ICU type ventilator.

DRIVING MECHANISMS

All ventilators require a driving force to deliver a gas flow into the patient. This force may either deliver the inspiratory gas directly to the patient or indirectly by compressing a bag or bellows containing the inspired gas mixture which in turn delivers the gasses to the patient.[2] Driving mechanisms which have been used in ventilators include the following:

1. Rotating electric motors. These may be used to:
   - **a.** drive a *crank* connected to a piston which in turn delivers the inspiratory gas. The disadvantage of these systems is that they have difficulty in altering inspiratory flow characteristics as the wheel and interlinked shaft drive will produce a fixed sinusoidal movement of the piston
   - **b.** power *low pressure blowers*. The advantage of this type of driving mechanism is that they can produce very high volumes of gas flow to enable the ability to cope with leaks from the patient circuit, in non-invasive ventilatory systems
   - **c.** drive *gas compressors* producing high pressure gas which may then be manipulated by pneumatic controls.
   - **d.** power a *piston* using a worm drive, this system is not seen in ITU ventilators (see Chapter 9, Dräger E anaesthetic ventilators).
2. Linear electric motors. These can be used to drive either pistons or diaphragms which are used in high frequency oscillators.
3. Tension springs which compress the gas in a storage bellows prior to being delivered to the patient. The disadvantage of this mechanism is that the pressure in the bellows is not constant but varies with the tension in the springs.
4. Weighted bellows. The gravitational force on a mass (the weight) will produce a constant driving pressure within the bellows; however, this mechanism is very susceptible to movement.

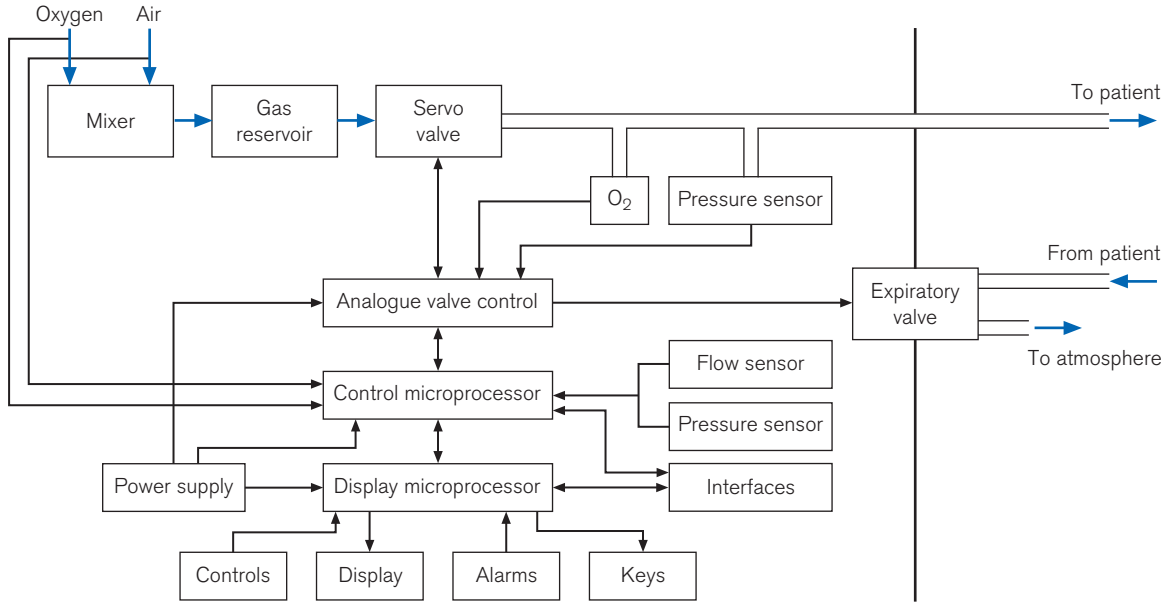


Figure 10.1 Microprocessor controls inside ventilator.

5. Pneumatic. These may be of two types:
   a. Low-pressure systems using gas supplied from electric blowers or venturi devices.
   b. High pressure with the gas supplied from a compressor via a high-pressure gas pipeline or cylinder which is connected to the ventilator.

MICROPROCESSOR ELECTRONIC CONTROL

The use of microprocessor control has become virtually ubiquitous in all modern ICU ventilators. At the heart of these is the control microprocessor whose function is to receive data from the analogue to digital converters that measure pressure and flow within the ventilator system. With this information and under software control, the microprocessor provides the commands for the circuit board to precisely control the inspiratory and expiratory flow valves and secondary functions (e.g. nebulizer control or to flush pressure measurement lines). Data are also received from the oxygen and carbon dioxide sensors, if fitted. Information may also be received from external sources and interface boards that allow ventilators to talk to each other for synchrony in ventilator modes such as for independent lung ventilation. At least one current device can use electromyographic input to aid in 'neural' control of the ventilator synchronisation, see below NAVA.

Some ICU ventilators use dual microprocessors to control the ventilator function. The advantage of using two processors in parallel is that not only can the function of information display be separated from the control of the valves, but each microprocessor can check the output of the other against its own computations to ensure maximum patient safety and ventilator reliability (Fig. 10.1).

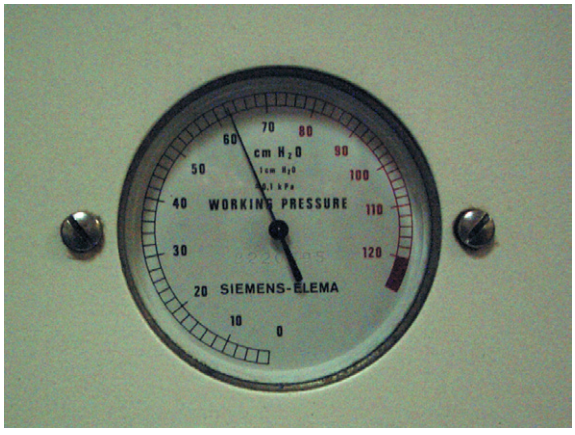


Figure 10.2 Analogue pressure display.

Information display

In the oldest ICU ventilators, the only information available to the operator was the analogue measurements of pressure or volumes (Fig. 10.2). With the same starting



Figure 10.3 LED digital display.

point of pressure measurement, microprocessor integration of the data now allows any number of derived variables to be monitored and displayed to inform the clinician of machine function and pulmonary mechanics. Alarm parameters can then be set for many of these values.

Digital information such as pressure values (peak, plateau, PEEP, etc.), calculations of tidal volume, machine or patient triggered breaths, inspired/expired volume differences and the quantity of gas leaking from the patient circuit were available to the user originally only in single-line numeric format (Fig. 10.3).[3] With the availability of liquid crystal display (LCD) screens, more of this information, instead of being provided in single numeric displays, is available on a matrix screen, allowing the user not only to see the static numerical data, but also graphical information, such as flow, pressure and volume variations. These may be plotted in parallel against time or as loops against one another. The increasing use of graphical displays allows the user to more easily understand the effect that a change in ventilator controls has on the delivery of gas to the patient. Although LCD screens have the advantage of low-power consumption, they are only readable over a narrow viewing angle and are in turn now being replaced by TFT LCD (thin film transistor LCD) screens that can provide similar information in colour. These are readable over a larger viewing angle and in lower lighting conditions (Fig. 10.4).



Figure 10.4 TFT panel display.

INSPIRATORY FLOW VALVE

Gas flow within ventilators may by controlled by one of two types of *solenoid valves*. In one type the valve has only two states: either being on or off; in the other type (*proportional valves*), the amount of opening when activated is directly related to the voltage applied to the solenoid.

High-speed proportional servo controlled (see below) flow valves are used in several manufacturers' ventilators. They are capable of delivering flows from 20 to 3000 ml s^{-1} and are adjusted by the ventilator's microprocessor control (servo control) using an electrodynamic motor similar to that of a loudspeaker. A current flowing in the field coils of the solenoid generates the force to move the piston up and down; connected to the piston is the valve orifice which opens to allow gas to flow. This type of valve has a

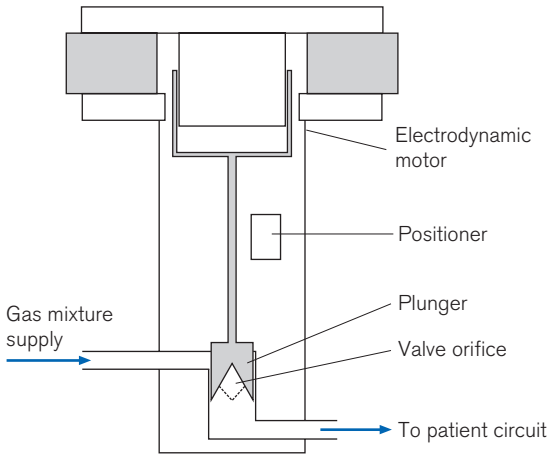


Figure 10.5 Inspiratory servo valve.

quick response time of typically 5 ms and with the small internal dead space of the ventilator, the flow rates change almost immediately in the patient circuit allowing precise control of the desired flow pattern and tidal volume. The microprocessor monitors the position of the valve and the pressure drop across it to enable it to continually adjust the flow to the required setting so that its performance characteristics are not affected by back pressure in the circuit (Fig. 10.5).

Some manufacturers use high-pressure, high-speed on-off gas solenoids, which control the flow of both oxygen and air. Under the microprocessor's control these can be rapidly pulsed on and off to create the desired inspiratory

their work of breathing. Together with the increased pressure changes, a time delay from the commencement of the patient's inspiratory effort to the start of gas flow will multiply the effect of any pressure drop; further increasing in the patient's work of breathing.[4]

Pressure triggering

Early ventilators such as the Siemens Servo 900C provided only pressure triggering; the pressure-sensing device being placed upstream within the inspiratory limb inside the ventilator. The pressure drop resulting from the patient's inspiratory effort activates the trigger whose sensitivity is usually set to −1–2 cm H_2O. When the threshold is reached, the electronic control within the ventilator opens the inspiratory valve allowing sufficient gas to reach the patient.[5] The disadvantage of this method of sensing is that due to the remote placement of the sensor the patient may have to generate a considerably greater effort at the patient end of the circuit to activate the trigger threshold within the ventilator. This was particularly apparent when thin-walled, highly compliant, disposable plastic tubing was connected to the ventilator. In addition if a hot water humidifier is also connected in the inspiratory limb, the added compliance and resistance of this further increases both the pressure gradient and the patient's work of breathing before obtaining any inspiratory gas supply.[6]

To overcome this problem, pressure sensing of the patient's inspiratory effort can be achieved more reliably in the expiratory side of the ventilator, although this is still subject to the effects of condensation accumulating within the expiratory limb of the patient circuit. Other manufacturers, such as Hamilton in their series of ventilators, sense pressure directly at the patient end of the circuit to avoid these problems.

Flow triggering

To overcome the disadvantages of pressure triggering most ICU ventilators now provide a form of flow triggering.[7] A flow trigger does not directly require change in pressure within the inspiratory circuit to enable a patient-initiated breath. The ventilators provide a continuous bias flow, frequently 10 l min^{-1}, which is introduced into the inspiratory circuit throughout all phases of the machine's respiratory cycle. When the patient starts to initiate a breath from the circuit, the fall in bias flow in the expiratory limb of the ventilator is sensed and the ventilator opens the inspiratory valve to allow a patient-initiated breath. The change in flow required to trigger a patient breath can be adjusted by the operator and is often set between 1 l min^{-1} and half the bias flow.[8] The advantage of this mode is that it is not as affected by humidifiers in the inspiratory limb or condensation in the tubing so patient work of breathing is minimized.[6,9,10] Most ventilators allow the user to adjust the sensitivity of the flow triggering to achieve the minimum work of breathing without the occurrence of falsely triggered breaths.[11]

Neurally adjusted ventilatory assist (NAVA)

Most ventilators rely on changes of either pressure or flow within the respiratory circuit to sense the patient's own breathing, but the Maquet Servo-i ventilator can use information from the patient's diaphragm. The electromyography signal from the diaphragm as the patient initiates a breath is sensed by an electrode array on a special nasogastric tube and these signals are used to synchronize the ventilator-assisted breaths more directly with the patient's own efforts.

EXPIRATORY PRESSURE GENERATION

The treatment of such conditions as adult respiratory distress syndrome (ARDS) requires the ventilators to be able to maintain a pressure in the patient circuit that is above atmospheric during expiration both for ventilator delivered breaths and for patient spontaneous respiration. PEEP/CPAP valves are incorporated into the expiratory limb and can be of two types: either fixed value or variable pressure.

- Fixed pressures can be generated by a spring-loaded valve, an underwater column or a weighted valve leaflet. The latter two types' function is severely impaired with movement and no longer used. It is worth noting that these types of valves exert a resistance throughout exhalation as a result of the mechanical force of the spring. Electronically controlled valves are usually only activated when the desired PEEP level is reached and do not impede exhalation until this point.
- Variable pressure valves have the advantage that they can be controlled (electronically) by the operator to produce the desired level of expiratory pressure and they can be closed completely by the ventilator to function as the expiratory valve as well.

Exhalation valves

Constriction type (scissor valves)

PEEP/CPAP can be produced using a variable orifice device in which the pinching of the expiratory rubber exit tube by an electromagnetic scissor valve can be used to control expiratory flow and pressure. During inspiration these valves can be closed completely to allow gas to flow into the patient's lungs (Fig. 10.9).

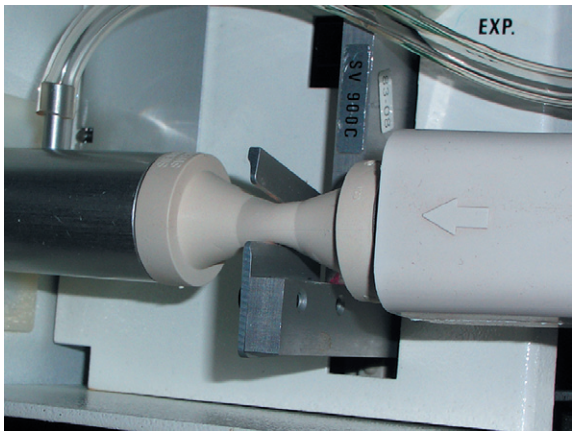


Figure 10.9 Scissor expiratory valve.

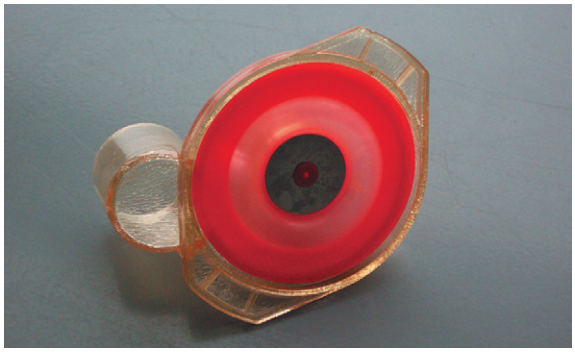


Figure 10.10 Expiratory valve diaphragm type.

Diaphragm type – mechanically operated

Levels of PEEP/CPAP can be generated using a linear electrodynamic motor that operates a large surface silicon membrane. Electrical current supplied to the motor during exhalation will produce a force on the diaphragm that is proportional to the level of PEEP/CPAP required. During inspiration the silicon diaphragm is forced hard onto the expiratory valve seat by the action of the actuator rod (Fig. 10.10). In normal exhalation the weight of the actuator shaft is balanced by a internal spring so the patient only has to overcome the weight of the membrane to exhale (Fig. 10.11); the expiratory resistance in this type of valve is less than 2 cm H_2O/L/s.[12]

Diaphragm type – pressure operated

The force on the expiratory valve diaphragm can be generated pneumatically within the ventilator in a similar fashion to that of the inspiratory gas flow, although from a single gas sources as it does not need to be a blend of air and oxygen. When this gas is applied to the non-patient side of the expiratory valve diaphragm it can be used to control the level of PEEP/CPAP in the ventilator circuit. The diaphragms used in these types of expiratory valve have a larger central mass (Fig. 10.12), to provide the required damping to prevent inadvertent oscillations.

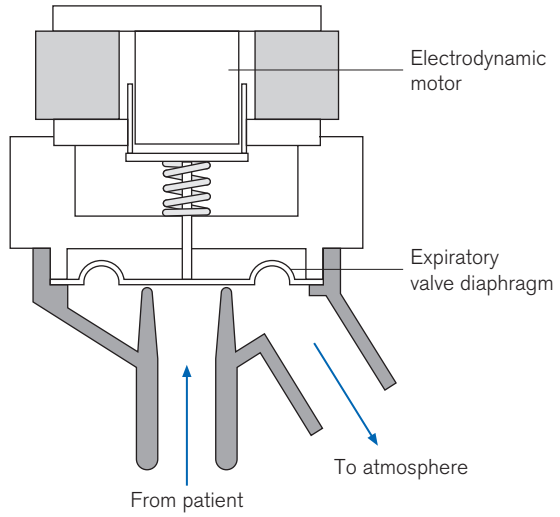


Figure 10.11 Expiratory valve.

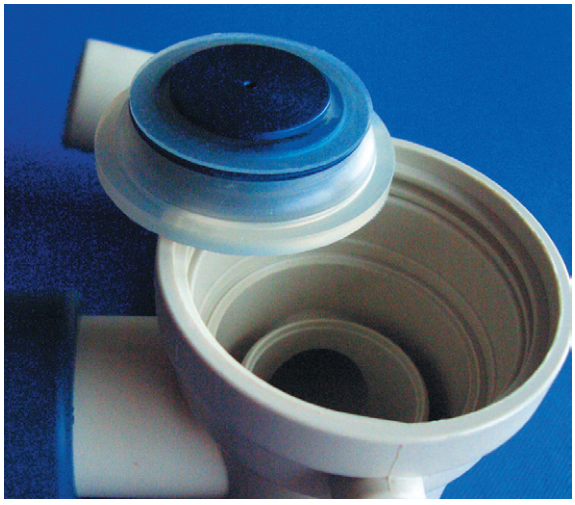


Figure 10.12 Expiratory valve Dräger Evita 4.

OVERPRESSURE VALVES

To ensure patients are not subjected to excessive airway pressures, ventilators provide protection in the form of an electronically controlled pressure limiter with a secondary mechanical device as back-up. The electronically controlled pressure limit can be set by the operator and this will



Figure 10.13 (From left to right) Auxiliary pressure, nebulizer and pneumotachograph on G5 ventilator (Hamilton Medical).

prevent pressure rising within the circuit even if the desired tidal volume is not delivered. Audible and visual alarms are triggered in any overpressure condition.

AMBIENT AIR INLET

In addition, for patient safety, an atmospheric air inlet valve is incorporated into the inspiratory gas pathway. In the event of a gas supply failure or internal electronic dysfunction these valves will open immediately allowing the patient to spontaneously inhale from ambient air.

NEBULIZER PORT

Ventilator manufacturers provide an additional gas source to drive a micro-nebulizer for the delivery of drug into the patient circuit (Fig. 10.13). The advantage of these fixed outlets is that this driving gas contains the same preset oxygen concentration as the respiratory gas, but it can be made to only be operational during the inspiratory phase so the drug is not wasted by being blown down the expiratory limb of the ventilator circuit. Unlike separate external gas flows these will not interfere with the ventilators' flow triggering sensitivity.

BATTERY BACK-UP

Microprocessor ventilators go though a series of computer self-checks to ensure that the electronic and pneumatics are working correctly on initial power-on; in some older models this checking process could take up to 40 s. To prevent this occurring in the event of a temporary disconnection of the power supply, ventilator manufacturers have provided a battery back-up to ensure the ventilator continues to function. 'Sleep' or 'suspend' buttons are also provided which enable the ventilator to stop operating, but then resume again with the same settings without the need to go through the initial self checks.

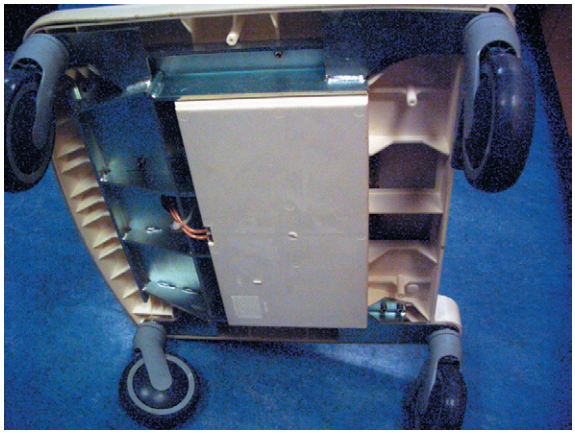


Figure 10.14 Battery back-up in base of ventilator.

Additional batteries can be provided to allow for more prolonged periods of disconnection from mains electrical supply to facilitate the movement of patients; these batteries are often stored in the base of the ventilator stand (Fig. 10.14). The addition of cylinder supplies of oxygen and air will allow the machine to be used as a transport ventilator.

FLOW PATTERN GENERATION/ VENTILATION MODES

With the use of sophisticated microprocessors and high-performance pneumatic controls, the gas flow characteristics of such ventilators are no longer primarily

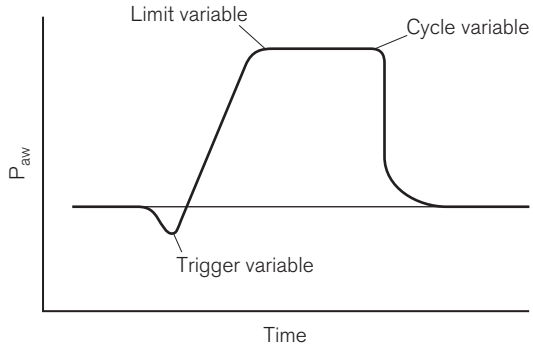


Figure 10.15 Inspiration showing position of phase variables.

determined by the physical characteristics of the machines, as had been the case with anaesthesia ventilators in the past.

The characteristics of the inspiratory and expiratory cycle can now be broken down into its fundamental components and gas delivery characteristics can be preset by the manufacturer or adjusted by the clinician at the bedside. Further advances in microprocessor monitoring are employed in some ventilators in turn allowing the patient's own breathing characteristics to alter ventilator setting, whose control traditionally had only been accessible to the care provider. This ability allows ventilators to operate in a closed loop automatic fashion.

A classification for mechanical ventilators proposed three types of variables:[13]

1. *Control* variables, which included pressure, volume, flow and rate
2. *Phase* variables, which defined how the change over points in the respiratory cycle occur, these being trigger, cycle and limit variables (Fig. 10.15)
3. *Conditional* variables, which define additional parameters, such as supplementary breaths and sighs.

Control and phase variables

When set up for a specific control mode, the ventilator delivers inspiratory gas flow to satisfy the appropriate limit of that variable; either a pre-set tidal volume in the *volume control mode* or a user defined airway pressure in the *pressure control mode*. Once initiated, inspiration occurs to the limit of that variable regardless of patient effort.

When a phase variable, such as time cycling (see below), is used in conjunction with the control modes (above) a more reliable delivery of appropriate minute ventilation ensues, especially if the patient is apnoeic or has limited respiratory effort. As the patient's respiratory function improves the control setting may be reduced to allow the patient to breathe alongside the minute ventilation delivered by ventilator. The control mode can be variably adjusted between full support and zero.

Volume pre-set control mode

In this mode, inhalation proceeds with a pre-set flow rate until the desired tidal volume is delivered. At the end of the predefined inspiratory time (the phase variable) passive exhalation then occurs. Inspiratory time can be set either directly or indirectly by adjusting either the ventilator rate and/or the inspiratory to expiratory ratio (I/E ratio). Inspiratory gas follows a predefined flow pattern and the peak pressure measured in the airways is a function of airways resistance and lung compliance (reflected in the force required for lung distension). Since the volume delivered is constant, peak airway pressure will alter with changing pulmonary compliance and airways resistance. However, true plateau pressure being a reflection of pulmonary system compliance only requires absence of gas movement and hence usually a degree of inspiratory hold to achieve a steady equilibrium. To avoid excessive pressures resulting in pulmonary barotrauma in a volume pre-set mode, most machines have high airway pressure alarms to alert the user to the potentially dangerous situation. In addition many machines have an overpressure release setting at which the ventilator will no longer deliver any additional tidal volume to the patient, but vents the excess to atmosphere. This facility may be governed by the high-pressure alarm limit or may be set independently and results in only partial delivery of the pre-set tidal volume. Warning alarms are triggered to alert staff to this situation.

When this mode of ventilation is used with hot water humidifiers in circuit, the rain out from these devices and subsequent pooling in the tubes may falsely trigger high-pressure alarms. A similar alarm may occur if the patient coughs during the inspiratory cycle.

Pressure pre-set control mode

In this mode, often referred to as pressure-controlled ventilation (PCV), a predefined inspiratory pressure is applied to the airways and the resulting pressure difference between the ventilator and the alveolus results in inflation until there is equilibrium between the two; after a pre-set inspiratory time, passive exhalation follows. In this mode the delivered volume during respiration is dependent on pulmonary and thoracic compliance. A potential disadvantage of this mode is that changes in pulmonary mechanics will result in varying tidal volumes and minute ventilation.[14] With accurate flow sensors within the circuit, close monitoring of tidal ventilation is employed and dramatic changes in tidal volume or minute ventilation will trigger a warning alarm.

As this mode of ventilation pre-sets the pressures within the inspiratory circuit, the flow pattern cannot be greatly influenced by the user, and is usually of a falling flow

pattern type as the pressure gradient between the inspiratory circuit and the alveolus declines with lung inflation. This usually results in a more homogenous gas distribution throughout the lung and improved arterial blood oxygenation and may reduce the patient's work of breathing compared to volume pre-set modes during assisted modes of ventilation.[15]

Other ventilator modes

The degree of sophistication of the new generation of ventilators has spawned an increase in the variety of and scope for ventilation strategies. This has led to an impressive increase in terminology to describe these. Unlike the nomenclature for drugs, there is no international standardization of the names attached to different modes of ventilation, even when they use similar control and phase variables throughout the respiratory cycle. Manufacturers have contributed to this confusion by using patents or trademarks to prevent similar names being used on other companies' machines (Table 10.1).

To fully understand what a ventilator will and will not allow a patient to do in each setting, there remains, sadly, the need to read the ventilator manual. For example: many modern ICU ventilators will allow spontaneous respiration, even during CMV mode (see below), and some will even apply pressure support to those breaths, whilst others will deliver only a machine volume or pressure pre-set breath on detecting an inspiratory effort. The principles of the more common terms are explained below.

Conditional variables

Controlled mandatory ventilation (CMV)

Breaths are delivered at pre-set time intervals regardless of patient effort. This mode is similar to that of anaesthesia ventilators and is most often used for the paralyzed or apnoeic patient. Large increases in work of breathing for the patient are created if their respiratory efforts fail to coincide with the ventilator's inspiratory cycle.

Intermittent mandatory ventilation (IMV)

In this mode breaths are delivered from the ventilator at pre-set intervals. However, patient's spontaneous respiration is allowed between ventilator-administered breaths.[16] The IMV rate may be reduced allowing increased time for the patient's spontaneous respiration during the weaning process.

Synchronous intermittent mandatory ventilation (SIMV)

In this mode the ventilator tries to deliver its breaths in conjunction with the respiratory effort of the patient. Spontaneous breathing is also allowed between ventilator-administered breaths (Fig. 10.16).

Synchronization of ventilator breaths is achieved by the ventilator attempting to detect the patient's inspiratory effort in a small time window prior to the initiation of its own inspiratory cycle. If patient inspiratory activity is detected the ventilator immediately delivers its mandatory breath. This mode of ventilation improves the comfort of spontaneous respiratory efforts for the patient and reduces the incidences of patient ventilator dyssynchrony where the patient appears to be 'fighting the ventilator'. Breath stacking may still occur where the patient wishes to exhale, but is then subject to an additional mandatory ventilatory inspiratory tidal volume, potentially over-distending the lungs.

On its own the SIMV mode has not been shown to improve patients' weaning from ventilatory support.[17] This may be due to the increased work of breathing associated with spontaneous respiration through these mechanical circuits. The work is created by having to generate sufficient pressures and flows within the ventilator tubing to trigger the opening of the ventilator's inspiratory valve for access to the extra gas flow required.

Pressure support mode/spontaneous assist

Pressure support (PS) ventilation has been shown to decrease the work of spontaneous breathing through ventilator circuits.[18,19] When triggered to do so, the ventilator produces a pressure in the respiratory circuit to support the patient's own inspiratory effort. The respiratory effort is detected either by flow or pressure triggering. With this mode of ventilation a user pre-set pressure is generated in the circuit (not a fixed tidal volume) to assist every patient spontaneous effort. This predefined airway pressure is sustained until the patient's own inspiratory flow falls below a predefined cut off, e.g. 25% of peak inspiratory flow[20] (Fig. 10.17). The disadvantage of this mode of ventilation is that if the patient fails to take any respiratory effort, no pressure supported breaths will be initiated. To avoid the potentially disastrous consequences most ventilators have a back-up apnoeic SIMV rate should the patient's spontaneous respiration cease.

For patients who have adequate respiratory drive and whose respiratory failure is not severe, PS ventilation may offer the patient considerable advantages, as all the breaths are patient initiated and breath stacking and fighting the ventilator are almost abolished. Even patients who are initially tachypnoeic, may be successful managed in this mode as the supporting pressure can be set sufficiently high to augment their own tidal volume and hence, reduce patient respiratory rate. As this is a pressure pre-set mode of ventilatory support, the risks of barotrauma associated with high airway pressures and fixed tidal volume ventilation are reduced.[21]

For patients who have severe respiratory failure this mode of ventilation is commonly used in conjunction with volume pre-set or pressure pre-set SIMV modes.[22]

Table 10.1 Nomenclature and ventilation strategies as termed by various manufacturers

ABBREVIATION	VENTILATORY MODE	MANUFACTURER	SIMILAR OR EQUIVALENT MODE
AC or ACV	Assist controlled ventilation		CMV, VC, VCV, IPPV
ACT	Automatic tube compensation		
APRV	Airway pressure release ventilation		
APVcmv	Controlled mandatory ventilation (adaptive pressure ventilation)	Hamilton Medical AG	
APVsimv	Synchronized intermittent mandatory ventilation (pressure regulated volume controlled)	Hamilton Medical AG	PRVC, PCV-VG, SIMV & Autoflow
ASB	Assisted spontaneous breathing		PS, PSV
ASV	Adaptive support ventilation (dual mode)	Hamilton Medical AG	
Autoflow and SIMV	SIMV plus autoflow (dual mode)	Dräger	APVsimv, PCV-VG, PRVC
Automode	Automode (dual mode)	Maquet	
BiLevel	Bi-level ventilation	Engstrom	DuoPAP, BiPAP, Bi-Vent, PCV+
BiLevel-VG	Bi-level ventilation volume guaranteed	Engstrom	
Bi-Vent	Bi-level ventilation	Maquet	DuoPAP, BiPAP, BiLevel, PCV+
BIPAP	Biphasic positive airway pressure	Dräger	DuoPAP, BiLevel, Bi-Vent, PCV+
BIPAP assist	Biphasic positive airway pressure	Dräger	DuoPAP, BiLevel, Bi-Vent, PCV+
CMV	Controlled mandatory ventilation		AC, ACV, VC, VCV, IPPV
(S)CMV (A/C)	Controlled mandatory ventilation (synchronized)	Hamilton Medical AG	AC,ACV, VC, VCV, IPPV
CPAP	Constant positive airway pressure		
DuoPAP	Bi-level ventilation	Hamilton Medical AG	BiPAP,Bivent, BiLevel, PCV+
HFJV	High-frequency jet ventilation		
HFOV	High-frequency oscillatory ventilation		
HFV	High-frequency ventilation		
ILV	Independent lung ventilation		
IPPV	Intermittent positive airway pressure ventilation		CMV, ACV
IRV	Inverse ratio ventilation		
MMV	Mandatory minute ventilation		
NAVA	Neurally adjusted ventilatory assist	Maquet	
NIV	Non-invasive ventilation		
P AC	Pressure assist controlled		PC
PAV	Proportional assist ventilation		
P-CMV (P-A/C)	Pressure controlled ventilation	Hamilton Medical AG	PCV, P AC
PCV or PC	Pressure controlled ventilation		P AC, P-CMV

Continued

Table 10.1 Nomenclature and ventilation strategies as termed by various manufacturers—cont'd

ABBREVIATION	VENTILATORY MODE	MANUFACTURER	SIMILAR OR EQUIVALENT MODE
PCV+	Biphasic positive airway pressure	Dräger	BIPAP
PCV-VG	Pressure controlled ventilation volume guaranteed	Engstrom	PRVC, APVsimv
PPS	Proportional pressure support		
PRVC	Pressure regulated volume control	Maquet	APVsimv, PCV-VG,SIMV & Autoflow
PS, PSV	Pressure support		ASB
P-SIMV	Synchronized intermittent mandatory ventilation (pressure controlled)	Hamilton Medical AG	
SIMV	Synchronized intermittent mandatory ventilation		
SIMV (PC)	Synchronized intermittent mandatory ventilation (pressure controlled)	Maquet	
SIMV (PRVC)	Synchronized intermittent mandatory ventilation (pressure regulated volume controlled)	Maquet	
SIMV (VC)	Synchronized intermittent mandatory ventilation (volume controlled)	Maquet	
SmartCare	SmartCare (dual mode)	Dräger	
SPONT	Spontaneous ventilation	Hamilton Medical AG	
VC	Volume control ventilation	Maquet, Dräger	CMV, AC,ACV
VCV	Volume control ventilation	Engstrom	CMV, AC, ACV
VS	Volume support	Maquet	

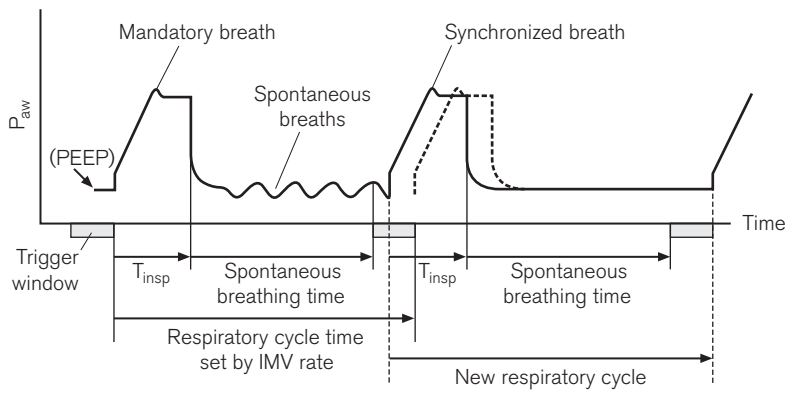


Figure 10.16 Synchronization of IMV breaths. The second tidal volume (synchronized breath) is delivered early because an inspiratory effort falls within the trigger window. The machine's next respiratory cycle timing is reset from this point.

Traditional 'control modes' of ventilation have only allowed the patient to breathe during the ventilator's expiratory phase. Most modern machines now provide a pressure preset control mode of ventilation that will allow the patient to breathe during any point of the ventilator respiratory cycle; although similar in function this mode is called by several names including BiLevel, Bi-Vent, BIPAP and DuoPAP.

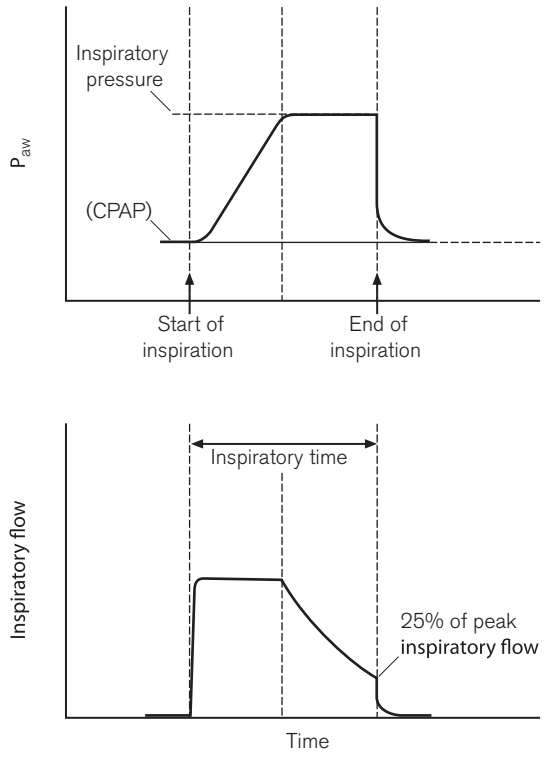


Figure 10.17 Pressure and flow curves when expiration trigger in pressure support mode is set to 25% of peak inspiratory flow.

Closed loop controlled ventilatory modes

Historically, adjustment of ventilator settings was made by physicians, nurses or attending bedside staff according to the patient's requirements. With the advent of microprocessor control it is possible to design ventilators with closed loop control, in which the ventilators themselves can adjust the way they deliver inspiratory gas flow.

The ability of ventilators to automatically wean clinically stable patients off ventilatory support has long been the aim of many manufacturers. Early ventilator models adjusted either tidal volume or inspiratory pressure to achieve a target end tidal carbon dioxide level. However, in intensive care patients, end tidal carbon dioxide does not always correlate with arterial carbon dioxide tension and the target variable of minute ventilation is more often used instead. Most ventilator manufacturers provide a closed loop mode of control implementing different versions of mandatory minute ventilation (MMV) (see Table 10.2) in which the target of minute ventilation is achieved by the ventilator adjusting inspiratory pressure and frequency.[23] Later versions of MMV have incorporated algorithms (Fig. 10.18) that aim to reduce the patient's work of breathing while still achieving the desired minute ventilation by encouraging the patient to breathe with larger tidal volumes and slower rates, this being more efficient than rapid shallow breathing.[24]

Dual control mode

Newer modes of ventilation referred to as dual control are where the ventilator acting in a closed loop feedback fashion measures parameters during the delivery of the breath and adapts its output in response to these changes according to a predefined algorithm.[3] These adaptations can be accomplished within a single breath when the ventilator changes from a pressure to a volume control mode. More commonly dual control breath-to-breath is used by ventilator manufacture to allow incremental changes to occur to achieve the desired outcome.

Table 10.2 Closed loop controlled ventilatory modes

NAME OF MODE	VENTILATOR	OPERATOR PRESET VARIABLES	VENTILATOR CONTROLLED VARIABLES
MMV and autoflow	Evita XL (Dräger)	MV, VT	F, Pinsp
Automode	Servo-i (Maquet)	VT,F	Pinsp, Psupp, Mode
ASV	G5 (Hamilton Medical)	%MV	Pinsp, Psupp, F, Ti

F, rate; VT, tidal volume; MV, minute volume; Pinsp, inspiratory pressure level; Psupp, pressure support level; Ti, Inspiratory time; MMV, mandatory minute ventilation; ASV, adaptive support ventilation

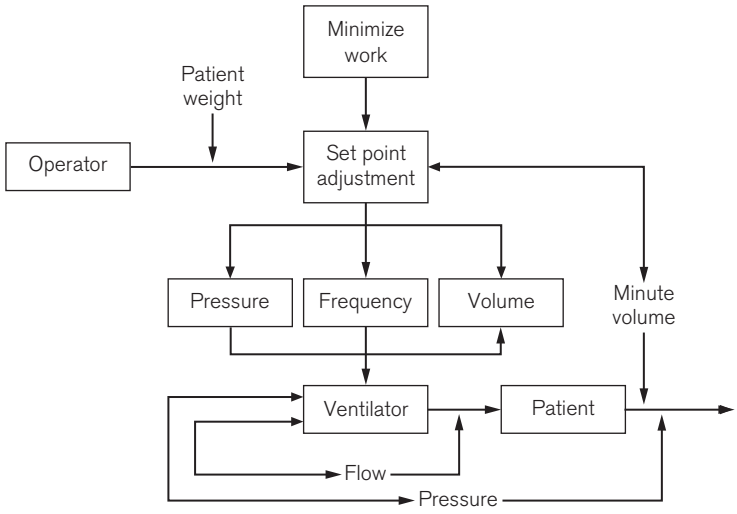


Figure 10.18 Algorithm used in adaptive support ventilation mode (ASV) on the Hamilton Galileo ventilator.

Volume support (VS)

VS is a mode where the ventilator acting in a pressure support mode either increases or decreases the applied airway pressure to ensure that the operator defined tidal volume is achieved.

Automode

This is a dual control mode in the Servo-I ventilator (Maquet), which supports the patient in a time cycled mode, a pressure regulated mode, a volume control mode or, if capable of spontaneous breathing, a flow cycled volume support mode.

Autoflow

In this mode the Dräger Evita ventilator changes from volume control to pressure control mode, adjusting the pressure to deliver the predefined tidal volume at the minimum pressure possible.

Adaptive support ventilation (ASV)

ASV is based on algorithm using the Otis equation[25] designed to minimize the patient's work of breathing by targeting tidal volume and respiratory rate. This dual mode works in either a pressure controlled or a pressure support mode, switching between the two modes depending on the patient's own respiratory activity.

Automatic tube compensation (ATC)

Ventilator manufacturers have tried to produce other additional modes for spontaneously breathing patients which attempt to reduce the resistance inherent with gas flows through an endotracheal tube. With the aim of keeping the pressure in the trachea at the desired level the ventilator adjusts inspiratory pressure according to the gas flow and direction.[26]

SmartCare

This dual control mode (found in some Dräger ITU ventilators) is designed to wean patients from ventilatory support who are capable of breathing in a pressure support mode by adjusting the level of pressure based on an algorithm with measures respiratory rate tidal volume end tidal carbon dioxide.

Total closed loop ventilation has not found universal favour as appropriate ventilation is not solely about carbon dioxide elimination and oxygenation of blood. It also involves complex interactions with the patient's neuromuscular effort, and that of their cardiovascular system.

Individual ventilators

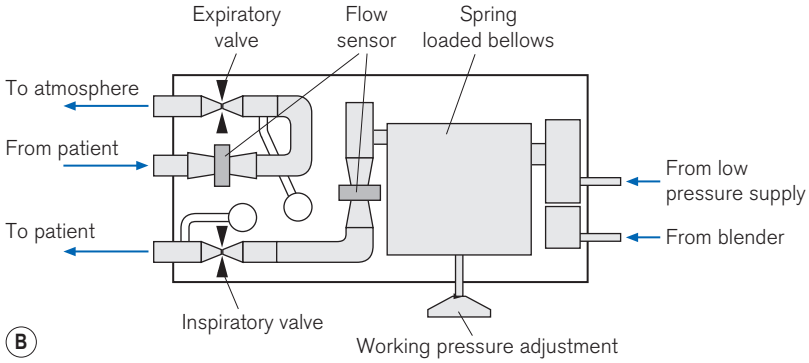
900C

The Siemens 900C ceased production in 2004 but is still in use and supported by the manufacturer. It has essentially two parts (Fig. 10.19): a pneumatic section sitting on top, and an electronic section below containing the ventilator controls and electronic displays. The two sections are connected by a cable and could be separated if desired. High-pressure gas enters the pneumatic section from an external blender. A second gas inlet connection is provided that accepts low pressure directly from an anaesthetic machine, if required. The gas then passes through an oxygen analyzer and main bacterial filter before entering a spring-loaded bellows, which stores the gas prior to use.

From the bellows, gas passes through the inspiratory flow transducer and then through the inspiratory scissor valve before leaving the unit to enter the patient circuit. Exhaled gas returning from the patient via the expiratory limb of the patient circuit re-enters the ventilator and passes through the expiratory flow transducer and expiratory scissor valve before exiting the unit via a one-way



A



B

Figure 10.19 **A.** Servo 900C series ventilator. **B.** Servo 900C working principles: diagram of the pneumatic section (see text).

valve to the atmosphere. The inspiratory scissor valve consists of a flexible piece of silicone rubber tubing that is compressed in the jaws of a scissor mechanism using an electric stepper motor to control gas flow. In contrast, the compression of the equivalent tube in the expiratory scissors valve is controlled and varied by a pull of an electromagnet under the control of the ventilator's electronics. This force is adjusted to maintain the correct PEEP in the expiratory limb; during inspiration the valve remains shut.

During inspiration, the gas flow is measured at the inspiratory flow transducer and compared in the electronics section to that which is required to achieve the operator's preset volume. If the actual flow does not match the required value, the stepper motor varies the compression of the inspiratory scissors valve to adjust the flow delivery. The driving pressure for the gas flow is generated by the pre-set tension in the spring attached to the inspiratory bellows, and during high inspiratory flow rates this may be insufficient to deliver the required gas flow. In this situation, the working pressure will have to be increased manually by the turning the key on the front of the pneumatic section (Fig. 10.20).



Figure 10.20 Servo 900C working pressure adjustment.

Servo-i and Servo 300

The Servo-i (Fig. 10.21A) and its predecessor the Servo 300 (last manufactured in 2003) has two units: a patient pneumatic unit and the control unit connected by a cable. The electronic circuit of the control unit both controls and displays the ventilator settings used to operate the pneumatic unit (Fig. 10.21B). Oxygen and air are supplied by pipeline and are blended directly into the patient's circuit by high-speed gas solenoid valves; unlike their predecessor (the 900C) there is no bellows storage for inspiratory gas and no low-pressure port for the supply of anaesthetic gas.

The solenoid valves have a response time of 6 ms under microprocessor control and can be rapidly opened or closed to achieve the desired flow rate and pattern in the ventilator circuit. In the Servo 300, the exhaled gas from the patient returns to the unit and passes through the expiratory flow transducer and pressure controlled expiratory valve before exiting out to the atmosphere. The Servo-i has a detachable expiratory cassette containing the entire expiratory gas flow pathway, together with the ultrasonic expiratory flow transducers and the expiratory valve. The expiratory pressure sensor and the actuator for the valve are housed in the body of the ventilator. The expiratory cassette can be autoclaved and the pressure sensor is protected by a bacterial filter.

A version of the Servo-i is manufactured in non-ferrous materials and the electronics shielded from the effect of high magnetic fields to allow it to be used to ventilate patients who are undergoing magnetic resonance imaging (MRI).

Dräger Evita series (2 Dura, 4, XL)

Dräger Evita 4 and XL series of ventilators have three sections; the *electronic compartment*, which sits directly on top of the *pneumatic controls,* and a third detachable *display unit*, which houses the controls and touch-sensitive screen (Fig. 10.22A). The screen displays both the ventilator information and the virtual touch sensitive buttons and dials. The 2 Dura model combines the display and the electronic and pneumatics in a single case.

Pneumatics (Fig 10.22B)

Gas from the high-pressure pipelines enters the ventilator via a filter and a non-return valve directly into the two proportional valves that control the flow of oxygen and air to be blended directly into the patient circuit. Sensors measure the pressure of the oxygen and air supplied to the ventilator and with this information a central microprocessor is able to adjust the function of the valves to deliver the correct flow into the patient circuit producing the inspiratory breath. The returning expiratory gas leaves the ventilator via a diaphragmatically operated expiratory valve and then through the external hot wire flow sensor finally to atmosphere. The PEEP/CPAP pressure is controlled by a balancing pressure from a separate proportional valve using the regulated oxygen supply applied to the downstream side of the expiratory diaphragm (see also Fig. 4.29). The gas flow of 9 l min^{-1} required to drive the nebulizer (when used) is also obtained from this oxygen supply. Here, the main two pneumatic valves are automatically re-adjusted to deliver the correct oxygen concentration.[27] During an oxygen pipeline failure, a switch-over valve is operated that allows the pneumatics to continue to function normally using the pressure from the air supply.

Non-invasive ventilation

The application of mechanical ventilatory support through a mask or helmet in place of endotracheal intubation is becoming increasingly accepted and utilized in the ICU. This modality of ventilatory support can be used successfully for patients with mild-to-moderate respiratory failure, but the patient must be mentally alert enough to follow commands, as without an endotracheal tube there is no mechanical method of preventing aspiration into the lungs. Clinical situations in which it has proven useful include acute exacerbation of chronic obstructive pulmonary disease (COPD) or asthma, and decompensated congestive heart failure (CHF) with mild-to-moderate pulmonary oedema. Conventional ICU ventilators set in their PSV mode of ventilation with PEEP are commonly use to support non-invasive ventilation through a mask,[4] although specially designed machines are now available for use in general acute wards (Fig. 10.23). Conventional ICU ventilators have the advantage of sophisticated monitoring and precise control of the oxygen concentration and inspiratory flow pattern, but set in their normal ventilatory modes they are ill adapted to cope with the large leaks of gas that may occur with poorly fitting masks and awake patients.[28] Manufacturers now often provide non-invasive



A

Figure 10.21 A. Servo-i.

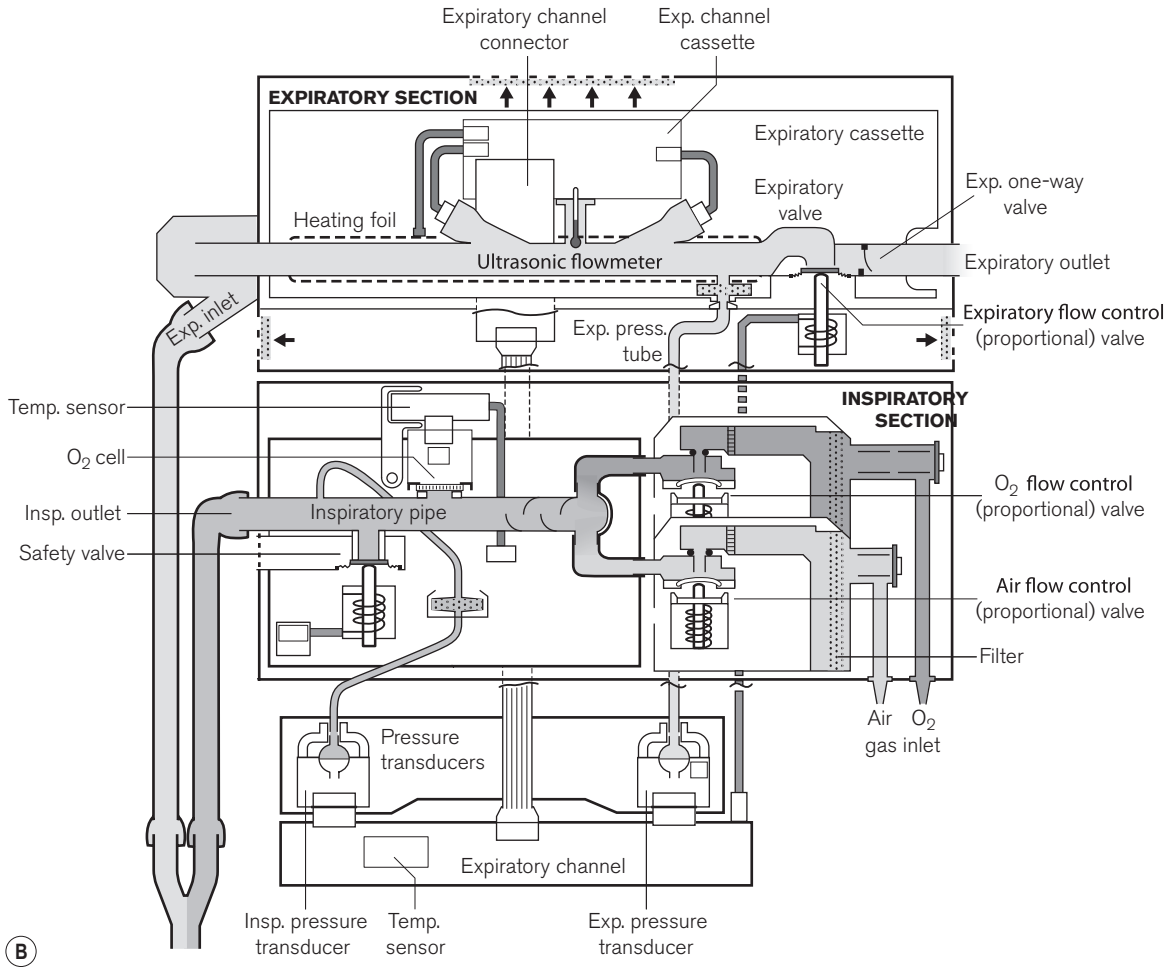


Figure 10.21, cont'd **B.** Working principles showing inspiratory section and detachable expiratory cassette.
Images courtesy of Maquet Critical Care.

modes on standard ICU ventilators that are more tolerant of the gas leakage from the patient circuit.

Specialist non-invasive machines use an electrically driven blower (Fig. 10.24) to deliver the inspiratory gas supply and are capable of generating flows of 300 L min^{-1}.[29] The rotating speed of the electrical blower is adjusted by the ventilator's microprocessor to achieve the operator's desired level of pressure for both inspiration and expiration. This bi-level pressure ventilatory mode is time cycled in a similar manner to the CMV mode on conventional ICU ventilator, the only difference being that patient is capable of breathing during the ventilator's inspiratory and expiratory cycle, as gas is supplied continuously during both phases of respiration. There is no expiratory valve in this type of ventilator, the gas leaving the system through a series of holes or slits close to the patient mask (Fig. 10.25A and B). Alterations in inspired oxygen concentration are achieved by entraining oxygen into the suction side of the blower or adding the oxygen flow directly into the patient circuit just before the mask; as a result it is difficult to achieve high inspired concentrations, particularly when the ventilation is delivering its maximum flow rate.

With the addition of a pneumotachograph in the inspiratory limb, by sensing changes of 40 ml s^{-1} in the flow required to maintain the expiratory pressure, sophisticated non-invasive ventilators enable the patient to trigger the commencement of the inspiratory positive airway pressure. Large and variable leaks from the mask make the detection of patient expiration by flow triggering difficult and, hence, less comfortable for the patient than a time cycled expiratory trigger.[30]

In the newer models of non-invasive ventilators manufacturers have incorporated pressure sensors and TFT

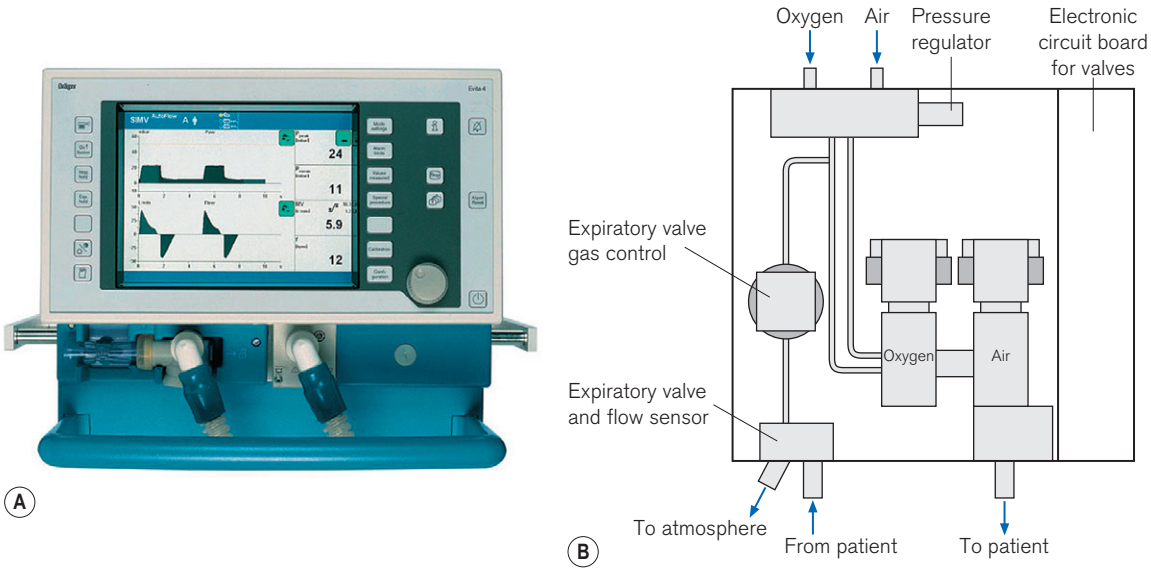


Figure 10.22 **A.** Dräger Evita XL. **B.** Dräger Evita 4 working principles.



Figure 10.23 Non-invasive ventilator BiPAP Focus.

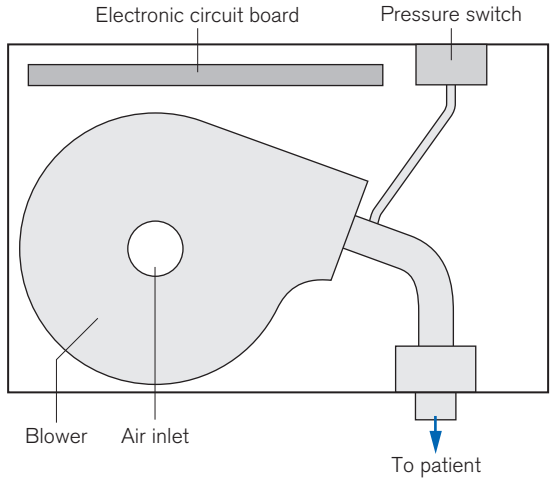


Figure 10.24 Working principles of a non-invasive ventilator.

matrix screens enabling the graphic display of pressure waveforms (Respironics BiPAP Vision). In addition, the fitting of oxygen sensors into the flow pathway allows more precise control of F_IO_2 and the ability to display the measured value on the screen.

High frequency oscillators

High frequency oscillators such as the SensorMedics 3100B or the Novalung Vision Alpha (Fig. 10.26A) use a device similar to a loudspeaker as the driving mechanism. The diaphragm (Fig. 10.26B) is made to operate at frequencies of 3–15 Hz or 180–900 breaths per minute, although typical starting settings are in the range of 5–6 Hz for adults.

A fresh gas supply or bias flow is provided constantly down the inspiratory limb independent of the diaphragmatic oscillations and can be controlled by the operator in a range of 20–60 L min^{-1} (Fig. 10.27). Altering the power setting (sometimes referred to as delta P) increases

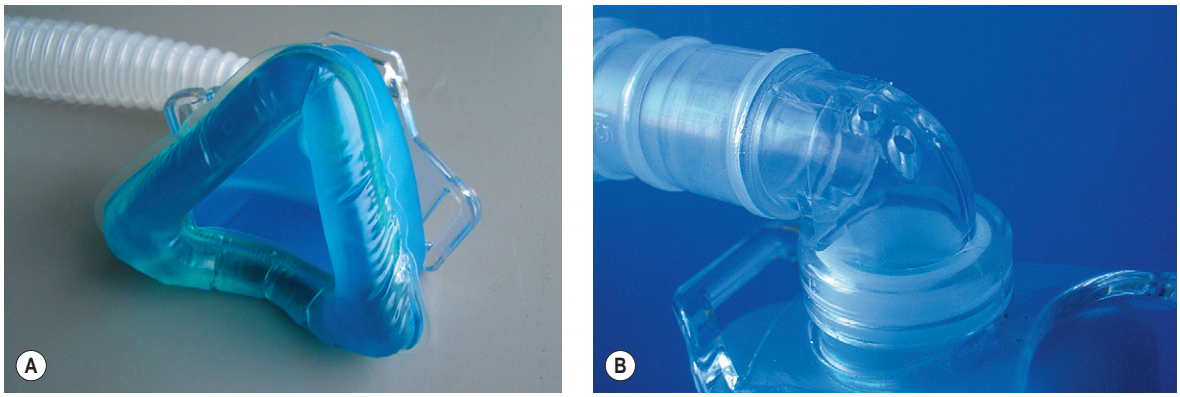


Figure 10.25 **A**. Mask for non invasive ventilation. **B**. Expiratory holes on mask mount for non-invasive ventilation.

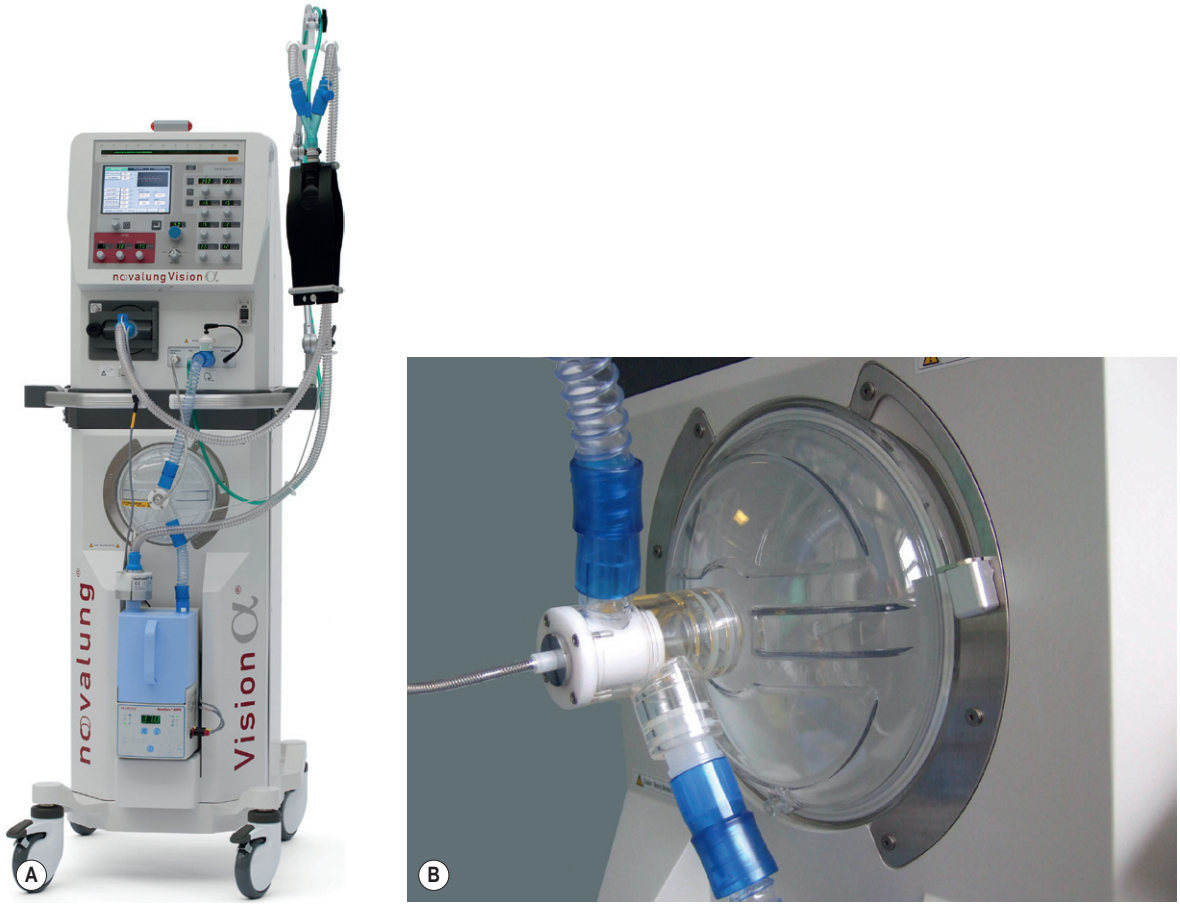


Figure 10.26 **A.** High-frequency oscillator ventilator. **B.** High frequency oscillator diaphragm. The white shuttle of a pneumatically operated valve termed the 'impedence valve' is visible: this isolates the diaphragm from the respiratory circuit when the ventilator is used in conventional mode.
Photograph courtesy of Inspiration Healthcare, UK.

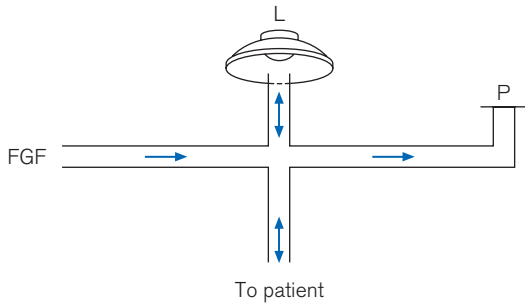


Figure 10.27 Schematic of high frequency oscillator ventilator. FGF, fresh gas flow; L, oscillator; P, PEEP valve.



Figure 10.28 Novalung, Interventional Lung Assist for extracorporeal CO_2 removal. The device is connected to two accompanying wide bore non-kink catheters inserted into the femoral vein and artery. Low pressure oxygen, from a standard flowmeter provides the sweep gas for CO_2 removal. Gas flow is through the hollow fibres of the membrane matrix.
Photograph courtesy of Inspiration Healthcare, UK.

the amplitude of the diaphragmatic oscillations. Carbon dioxide elimination from the patient is achieved by a combination of altering the power setting or increasing the bias flow.[31] The inspiratory proportion of the total time can also be adjusted by the operator. Alternatively the CO_2 removal may be largely accomplished by extrapulmonary means using simple, pumpless, miniature extracorporeal circuits (termed 'interventional lung assist' devices) (Fig. 10.28), allowing even greater degrees of lung rest.[32,33]

The Vision Alpha can also be used as a conventional ventilator in a CMV or CPAP mode. Humidification of patient gasses is achieved by a conventional-type water bath arrangement (the blue single-use patient canister can be seen in Fig. 10.26A) and to prevent rain out, the spiral anti-kink bindings of the patient circuit are embedded throughout with a heating wire. Low compliance 15 mm diameter tubing is used.

SUMMARY

Despite the ideals that one machine should cope with all clinical situations, modern ICU ventilatory strategies require a range of ventilators with different operational modalities to facilitate the full range of requirements. Safe and easy-to-use machines are important for clinical practice with the ability to ventilate while not restricting the patient's spontaneous breathing.

REFERENCES

1. Del Valle RM, Hecker RB. A review of ventilatory modalities used in the intensive care unit. *Am J Anesthesiol* 1995;**22**:23–30.
2. Smallwood RW. Ventilators – reported classifications and their usefulness. *Anaesth Intensive Care* 1986;**14**:251–7.
3. Chatburn RL. Computer control of mechanical ventilation. *Respir Care* 2004;**49**:507–17
4. Nava S, Ambrosino N, Bruschi C, Confalonieri M, Rampulla C. Physiological effects of flow and pressure triggering during non-invasive mechanical ventilation in patients with chronic obstructive pulmonary disease. *Thorax* 1997;**52**: 249–54.
5. Sassoon CS, Gruer SE. Characteristics of the ventilator pressure- and flow-trigger variables. *Intensive Care Med* 1995;**21**:159–68.
6. Street MK, Hopkinson RB. Evaluation of the comfort of spontaneous respiration through three ventilator systems. *Intensive Care Med* 1987;**13**:405–10.
7. Prinianakis G, Kondili E, Georgopoulos D. Effects of the flow waveform method of triggering and cycling on patient-ventilator interaction during pressure support. *Intensive Care Med* 2003;**29**:1950–9.
8. Sassoon CS. Mechanical ventilator design and function: the trigger variable. *Respir Care* 1992;**37**: 1056–69.
9. Giuliani R, Mascia L, Recchia F, Caracciolo A, Fiore T, Ranieri VM. Patient-ventilator interaction during synchronized intermittent mandatory ventilation. Effects of flow triggering. *Am J Respir Crit Care Med* 1995;**151**:1–9.
10. Ranieri VM, Mascia L, Petruzzelli V, Bruno F, Brienza A, Giuliani R. Inspiratory effort and measurement of dynamic intrinsic PEEP in COPD patients: effects of ventilator triggering systems. *Intensive Care Med* 1995;**21**:896–903.

11. Imanaka H, Nishimura M, Takeuchi M, Kimball WR, Yahagi N, Kumon K. Autotriggering caused by cardiogenic oscillation during flow-triggered mechanical ventilation. *Crit Care Med* 2000;**28**:402–7.
12. Techical Specification. *Veolar operators manual*. Rhazuns: Hamilton Medical, 1992.
13. Chatburn RL. Classification of mechanical ventilators. *Respir Care* 1992;**37**:1009–25.
14. Kallet RH, Alonso JA, Diaz M, Campbell AR, Mackersie RC, Katz JA. The effects of tidal volume demand on work of breathing during simulated lung-protective ventilation. *Respir Care* 2002;**47**:898–909.
15. Kallet RH, Campbell AR, Alonso JA, Morabito DJ, Mackersie RC. The effects of pressure control versus volume control assisted ventilation on patient work of breathing in acute lung injury and acute respiratory distress syndrome. *Respir Care* 2000; **45**:1085–96.
16. Sassoon CS. Intermittant mandatory ventilation. In: Tobin MJ, editors. *Principles and practice of mechanical ventilation*. New York: McGraw-Hill; 1194. p. 221–37.
17. Brochard L, Rauss A, Benito S, Conti G, Mancebo J, Rekik N, et al. Comparison of three methods of gradual withdrawal from ventilatory support during weaning from mechanical ventilation. *Am J Respir Crit Care Med* 1994;**150**:896–903.
18. Brochard L, Harf A, Lorino H, Lemaire F. Inspiratory pressure support prevents diaphragmatic fatigue during weaning from mechanical ventilation. *Am Rev Respir Dis* 1989;**139**:513–21
19. Esteban A, Frutos F, Tobin MJ, Alía I, Solsona JF, Valverdú I. A comparison of four methods of weaning patients from mechanical ventilation. Spanish Lung Failure Collaborative Group. *NEJM* 1995; **332**:345–50.
20. Du HL, Amato MB, Yamada Y. Automation of expiratory trigger sensitivity in pressure support ventilation. *Respir Care Clin N Am* 2001;**7**:503–17, x.
21. Moylan FM, Walker AM, Kramer SS, Todres ID, Shannon DC. The relationship of bronchopulmonary dysplasia to the occurrence of alveolar rupture during positive pressure ventilation. *Crit Care Med* 1978;**6**:140–2.
22. Esteban A, Anzueto A, Alia I, Gordo F, Apezteguía C, Pálizas F, et al. How is mechanical ventilation employed in the intensive care unit? An international utilization review. *Am J Respir Crit Care Med* 2000;**161**:1450–8.
23. Brunner JX. Principles and history of closed-loop controlled ventilation. *Respir Care Clin N Am* 2001;**7**:341–62, vii.
24. Brunner JX, Iotti GA. Adaptive Support Ventilation (ASV). *Crit Care Med* 2002;**68**:365–8.
25. Otis AB, Fenn WO, Rahn H. Mechanics of breathing in man. *J Appl Physiol* 1950;**2**:592–607.
26. Guttman J, Erbhard L, Fabry B, Bertschmann W, Wolff G. Continuous calculation of intratracheal pressure in tracheal intubated patients. *Anesthesiology* 1993;**79**:503–13.
27. Service Manual Evita 4. 5th ed. Lubeck: Dräger Medizintechnik GmbH; 2000.
28. Mehta S, McCool FD, Hill NS. Leak compensation in positive pressure ventilators: a lung model study. *Eur Respir J* 2001;**17**:259–67.
29. BiPAP Vision: Respironics, 2003.
30. Calderini E, Confalonieri M, Puccio PG, Francavilla N, Stella L, Gregoretti C. Patient-ventilator asynchrony during noninvasive ventilation: the role of expiratory trigger. *Intensive Care Med* 1999;**25**:662–7.
31. Fort P, Farmer C, Westerman J, Johannigman J, Beninati W, Dolan S, et al. High-frequency oscillatory ventilation for adult respiratory distress syndrome – a pilot study. *Crit Care Med* 1997;**25**:937–47.
32. David M, Heinrichs W. High-frequency oscillatory ventilation and an interventional lung assist device to treat hypoxaemia and hypercapnia. *Br J Anaesth* 2004;**93**:582–6.
33. Lubnow M, Luchner A, Philipp A, Buchner S, Jeron A, Karagiannidis C, et al. Combination of high frequency oscillatory ventilation and interventional lung assist in severe acute respiratory distress syndrome. *J Crit Care* 2010 Sept;**3**:436–44.

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Chapter 11

Breathing filters, humidifiers and nebulizers

Antony R Wilkes

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Most patients undergoing surgery and those in intensive care have an airway device *in situ*. This allows the delivery of various gas mixtures and/or vapours, and any necessary ventilatory assistance. In addition, a tracheostomy may be carried out on some patients to bypass the upper airways, either temporarily or for the longer term, whilst also allowing the patient to speak, eat and drink. These devices bypass the normal physiological functions of the nasopharynx.

During normal breathing, the nasopharynx warms, humidifies and, particularly during nasal breathing, filters inspired gasses. When the patient's nasopharynx is bypassed, these functions are lost. The trachea has a continuous stream of mucus, called the mucociliary elevator; this moves towards the pharynx, trapping and removing any particles that enter the trachea. The mucociliary elevator relies on optimum levels of temperature, and particularly humidification, to work effectively.

Gasses supplied from cylinders or pipelines need to be very dry to reduce the risk of corrosion, condensation and frost forming in cylinders, pipes and valves. Gasses delivered to the patient's trachea, therefore, need to be artificially *warmed*, *humidified* and *filtered* to prevent damage to the patient's airways,[1,2] to maintain the effectiveness of the mucociliary elevator[3] and to reduce the incidence of infection.[4] This chapter deals with devices that fulfill these functions.

BREATHING SYSTEM FILTERS

Filtration and mechanisms of filtration

Filtration is the removal of particles from either a gas or a liquid suspension. Filters are used to remove particles from gasses delivered to patients, to prevent microbes from patients cross-infecting other patients and staff, and to reduce the contamination of equipment. In addition, sputum expectorated by a patient and condensation in breathing systems may harbour pathogens, and filters can be used to reduce the risk of liquid-borne cross-infection.[5]

Mechanisms of filtration of gas-borne particles

Filter material generally consists of fibres formed into a non-woven wad or sheet. There are five main mechanisms by which the filter material removes particles from a flow of gas (Fig. 11.1):

a. Interception

Particles will tend to follow streamlines in a flow of gas. However, if a particle in the gas stream comes within one particle radius of the surface of a fibre, the particle will adhere to the fibre.

b. Inertial impaction

Particles have mass and are, therefore, not always able to follow a gas streamline around a fibre due to their

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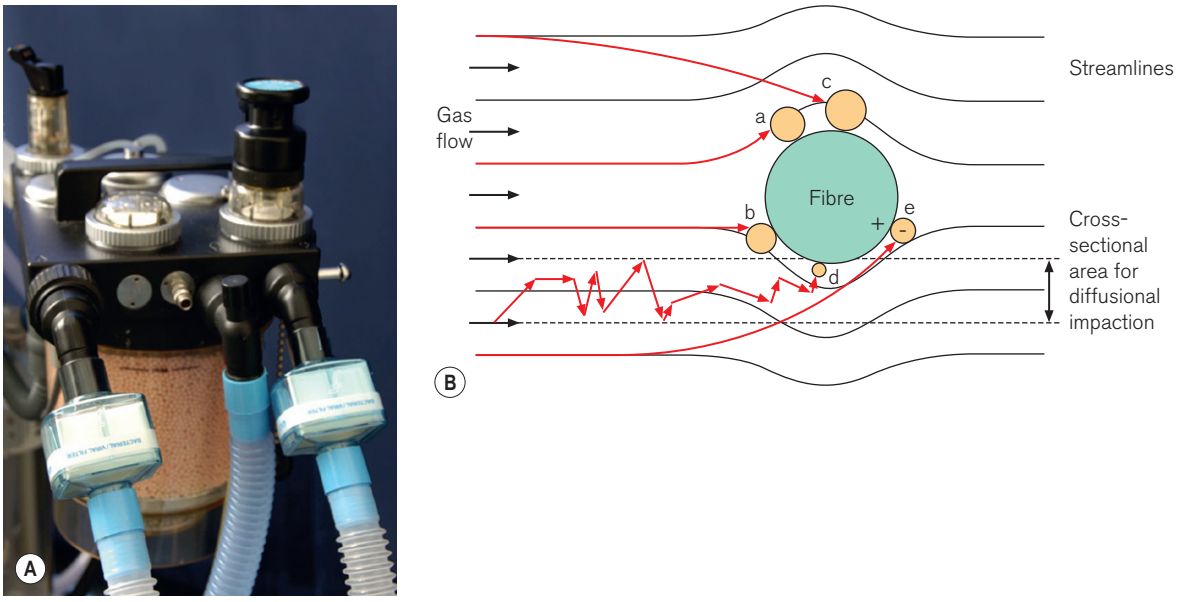


Figure 11.1 **A.** Filters on absorber block. **B.** Mechanisms of filtration. (a) Interception; (b) inertial impaction; (c) gravitational settling; (d) diffusional impaction; (e) electrostatic attraction (see text for details).
Adapted from Hinds WC. Aerosol Technology. Properties, behaviour, and measurement of airborne particles. *2nd ed. New York: John Wiley and Sons; 1999.*

inertia. The particles may, therefore, strike the fibre, even though the gas streamline is more than one particle radius from the fibre.

c. Gravitational settling

Large particles in slow-moving air do not follow gas streamlines, but settle due to gravity, and can, therefore, fall onto, and adhere to, a fibre.

d. Diffusion

Small particles do not remain on particular streamlines in the gas but undergo Brownian motion due to interactions with gas molecules. This effectively increases their cross-sectional area and so increases the probability of them striking a fibre.

e. Electrostatic attraction

Some filter material (see below) is electrostatically charged during manufacture to enhance its ability to capture particles. There are three mechanisms of capture: charged particles in the gas stream are attracted to oppositely charged fibres; neutral particles are attracted to a charged fibre as the electric field on the fibre induces a dipole in the particles (positive and negative charges on opposite sides of the particles); and charged particles are attracted to neutral fibres by inducing image forces on the fibres.

Most penetrating particle size

The relative efficiencies of the five mechanisms of filtration vary with the size of the particle (Fig. 11.2). In particular, particles of a certain size, typically in the range 0.05–0.5 µm, pass through the filter more easily than others. This size is known as the most penetrating particle size. Particles of this size are too small to be directly intercepted by fibres and too large to undergo substantial Brownian motion.

Types of filter

There are two main types of filter material used in breathing system filters:

Glass fibre filters

This filter material consists of a sheet of resin-bonded glass fibres. The fibres are packed densely (Fig. 11.3A) and hence the sheet has a high resistance to gas flow per unit area. A sheet with a large surface area is used to reduce the resistance to gas flow to an acceptable level. The sheet is then pleated to minimize the required volume, and hence dead space, for the housing. This type of filter material is *hydrophobic* and under normal clinical conditions, does not absorb water.

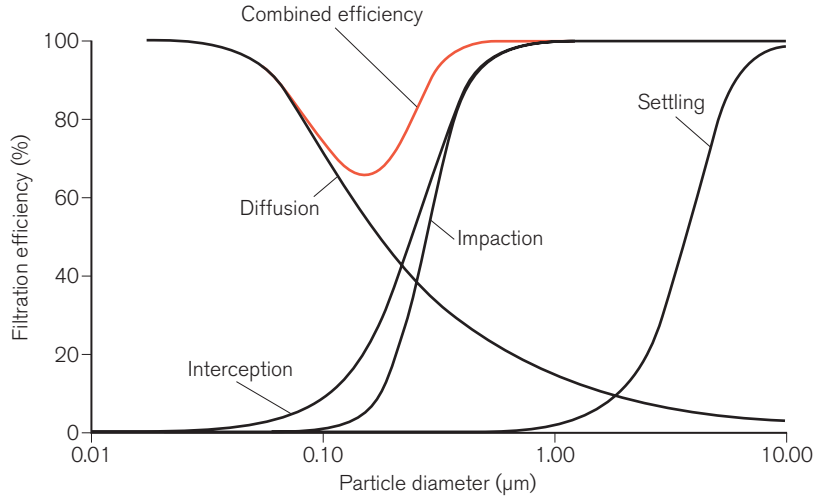


Figure 11.2 Filtration efficiency as a function of particle aerodynamic diameter due to the different filtration mechanisms. Note the minimum efficiency (maximum penetration) at a diameter of about 0.3 μm. *Adapted from Hinds WC. Aerosol Technology.* Properties, behaviour, and measurement of airborne particles. *2nd ed. New York: John Wiley and Sons; 1999.*

Electrostatic filters

There are two main types of electrostatic filter material. In both types, the fibre density is lower than in glass fibre filters and hence the resistance to gas flow is lower per unit area. The filtration performance is enhanced by using electrostatically charged material, which attracts and binds with any particles passing through the filter material. Therefore, this type of filter material does not need to be pleated, and a flat layer is generally used in breathing system filters.

1. Tribocharged filters

An electrostatic charge can be induced on two dissimilar fibres by rubbing them together during the manufacturing process, so that one type becomes positively charged and the other type negatively charged (*tribocharging*). One such filter material is made from fibres of polypropylene and modacrylic which can then be converted into a non-woven felt (Fig. 11.3B).

2. Fibrillated coronal-charged filters

An electrostatic charge can be applied to a sheet of polypropylene by using a point electrode emitting ions (*corona charging*). An opposite charge can be induced on the rear of the sheet. This type of material is often called an electret. If the sheet of polypropylene is now stretched, the strength of the molecular bonds is enhanced in the direction of the stretching, but reduced in a direction perpendicular to it. The sheet can then be split into fibres – a process called fibrillation – and made into a non-woven filter wad (Fig. 11.3C).

Measuring the performance of breathing system filters

The filtration efficiency of a filter is determined by measuring the number of particles passing through the filter as a percentage of the number of particles in an aerosol challenge to the filter. This percentage is the penetration value for the filter. Although challenges of microbes can be used,[6] the standard for breathing system filters specifies that the challenge should consist of a particular quantity of an aerosol containing sodium chloride particles having diameters close to the most penetrating particle size.[7] The filter is challenged at a flow of 15 or 30 L min^{-1} for filters intended for use with paediatric or adult patients, respectively. Typical penetration values for filters are shown in Fig. 11.4.

HUMIDIFIERS

Humidity

Humidity is used to describe the amount of water vapour in air or gas. The mass of water vapour in the gas is the absolute humidity (g m^{-3}). The maximum amount of humidity that gas can contain is limited by temperature (Fig. 11.5). At the maximum humidity for a particular temperature, the gas is said to be saturated with water vapour, and the level of humidity is the humidity at saturation. The relative humidity (RH; %) is the absolute humidity of the gas at a particular temperature as a percentage of the humidity at saturation at the same temperature.

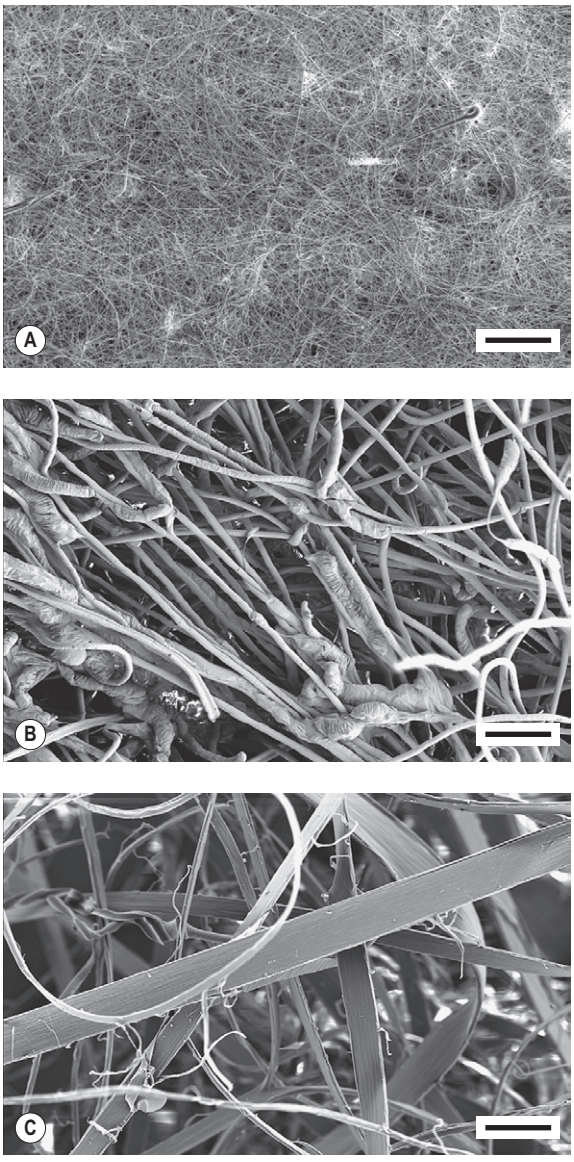


Figure 11.3 Scanning electron microscope photographs of the surface of different types of filter material. **A.** Glass fibre; **B.** tribocharged electrostatic; **C.** fibrillated electrostatic. Note the 100 µm scale marker in the bottom right corner of the photographs. Bacteria are about 1 µm in diameter.

Room air at 22°C typically has an absolute humidity level of approximately 10 g m^{-3}. The humidity at saturation of air at 22°C is approximately 20 g m^{-3}, so that the room air has a relative humidity of approximately 50% RH. If the air is cooled, a point is reached at about 11°C when the absolute humidity level equals the humidity at saturation, and hence the relative humidity is 100%. If the air cools to an even lower temperature, condensation will occur. If room air is inspired, the air is warmed to 37°C by the upper air passages by the time it reaches the lungs, and the humidity is increased from 10 to 44 g m^{-3} (BTPS conditions: body temperature and pressure, saturated). The difference between the two (−34 g m^{-3}) is the humidity deficit: humidity must be added by the airways to reduce this deficit to 0 g m^{-3}. If the room air is warmed from 22 to 37°C without any humidification, the relative humidity will fall to 100 × (10 ÷ 44) = 23%. The massic enthalpy (latent heat) of evaporation of water is 2.4 kJ g^{-1}. To saturate inspired gasses, which have a low level of humidity, a considerable proportion of the body's heat production must be used (up to one-third for a neonate). This can then lead to a fall in the patient's core temperature of more than 1°C.

Humidification requirements

The level of humidity acceptable in gasses delivered to patients whose upper airways have been bypassed depends on the length of time of the bypass. For short-term use the level of humidity may only need to be 20 g m^{-3} (45% of BTPS conditions).[8] For longer use, for example, for patients in intensive care units, the level of humidity should be at least 33 g m^{-3} (75% of BTPS conditions).[9] One group has proposed that the level of humidity should be close to BTPS conditions (44 g m^{-3}).[3]

Humidification equipment

Gas can be humidified using either passive or active systems. Passive systems, such as heat and moisture exchangers (HMEs), rely on the patient's ability to add moisture to the inspiratory gas. Active systems, such as heated humidifiers, add water vapour to a flow of gas independently of the patient. Combined passive and active devices are also now available.

Passive humidification systems

1. Heat and moisture exchangers (HMEs)

HMEs return a portion of the exhaled heat and moisture to the next inspiration. If the exhaled gas is at 34°C and is saturated with water vapour (38 g m^{-3}) then, even if the HME is 80% efficient, only 30 g m^{-3} is returned to the patient.

These devices generally consist of a transparent plastic housing so that any obstructions and secretions in the device can be seen readily. The housing contains a layer of either foam or paper that is commonly coated with a hygroscopic salt such as calcium chloride. The expired gas cools as it passes through this layer and condensation occurs, releasing the massic enthalpy of vaporization, which is partly retained by the HME layer. The hygroscopic salt absorbs water vapour, hence reducing the relative humidity of the gas to below saturation level, although

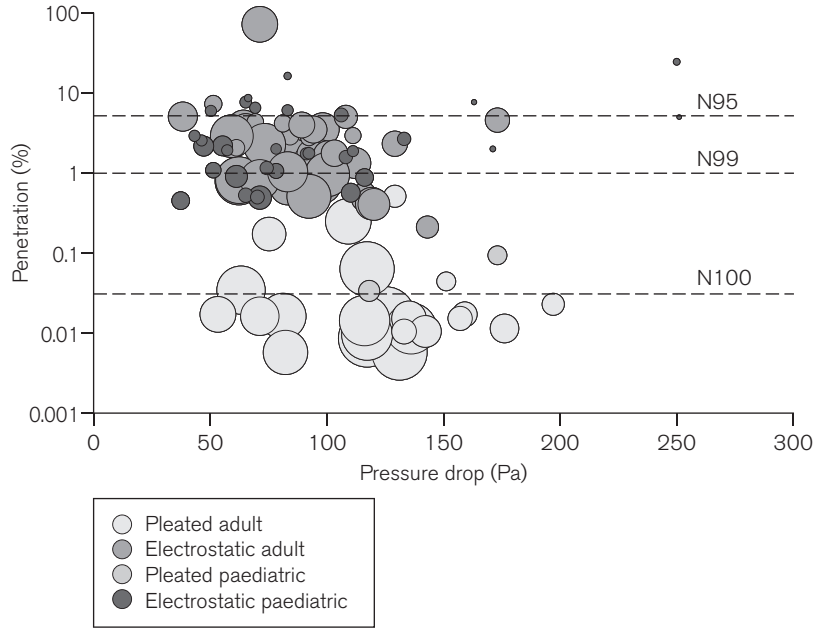


Figure 11.4 Penetration through 104 different filters when challenged with an aerosol of sodium chloride particles at the most penetrating particle size against pressure drop measured across the filter (100 Pa ≈ 1 cm H_2O). Filters were tested at flows of 15 l min^{-1} (paediatric) or 30 l min^{-1} (adult), respectively. The size of each bubble is proportional to the internal volume of the filter. An ideal filter would have low values for penetration, pressure drop and internal volume (a small 'bubble' in the lower left hand side of the figure). However, filters with low penetration (high filtration efficiency) and low-pressure drop tend to be large filters; smaller filters tend to have higher pressure drops and higher penetration values (low filtration performance). N95, N99 and N100 refer to filtration efficiencies: namely <5%, <1% and <0.03% penetration respectively.
Adapted from Wilkes AR. Heat and moisture exchangers and breathing system filters: their use in anaesthesia and intensive care. Part 1 – history, principles and efficiency. Anaesthesia *2011;**66**:31-39 with permission from Wiley-Blackwell.*

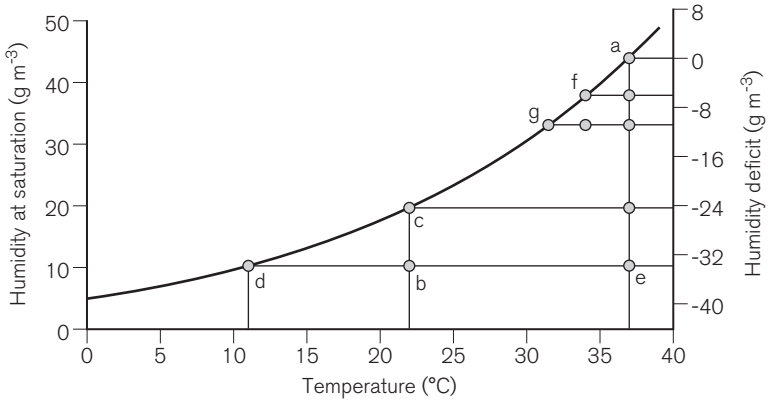


Figure 11.5 Humidity at saturation against temperature. Humidity deficit indicates the humidity that must be added by the airways to increase the humidity to (a) BTPS conditions (37°C, 44 g m^{-3}, 100% relative humidity). (b) Typical room air (22°C, 10 g m^{-3}, 50% RH); (c) room air saturated with water vapour (22°C, 20 g m^{-3}, 100% RH); (d) room air cooled so that condensation occurs (11°C, 10 g m^{-3}, 100% RH); (e) room air warmed to body temperature (37°C, 10 g m^{-3}, 23% RH); (f) expired gas (34°C, 38 g m^{-3}, 100% RH); (g) minimum moisture output for humidifiers intended for use with patients whose airways have been bypassed (33 g m^{-3}).

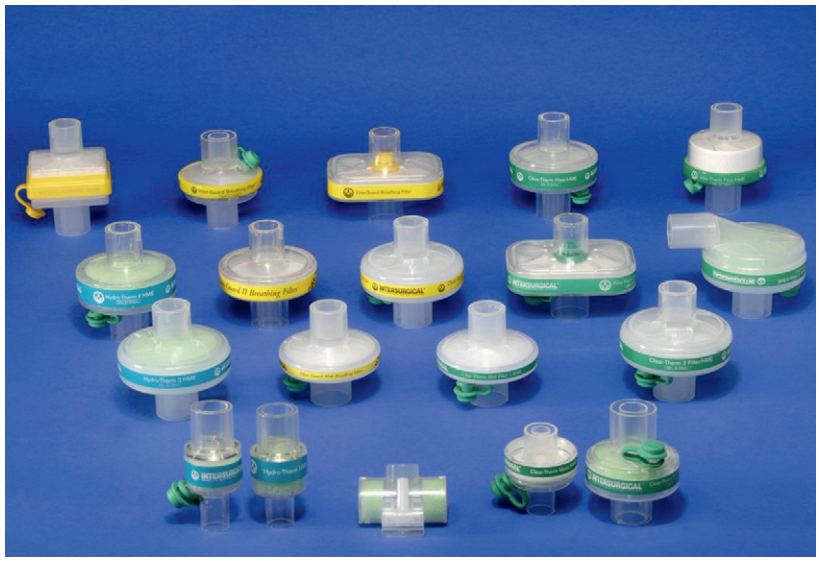


Figure 11.8 Examples of HMEs, filters and HMEFs from one manufacturer. Note the colour-coding: blue, HME-only; yellow, filter-only; green, HMEF. Other manufacturers use different colour-coding. The different sizes of devices are intended for use with paediatric and adult patients.

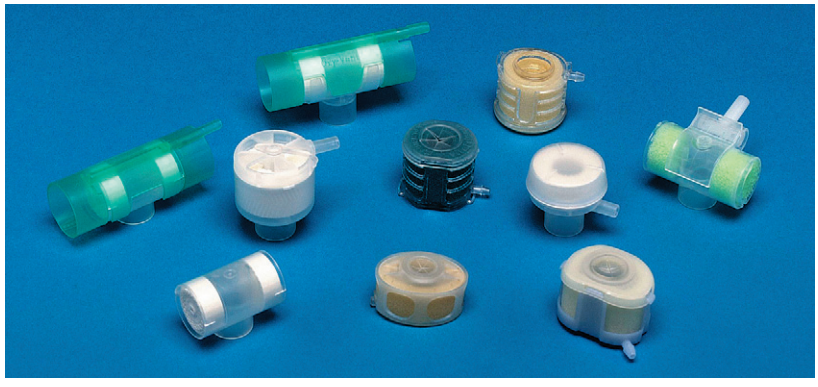


Figure 11.9 Heat and moisture exchangers intended to be attached to a tracheostomy tube. An oxygen tube can be connected to a port on some devices; some devices also have a port through which the patient's airways can be suctioned. One device has a valve which, when pressed closed, occludes, allowing the patient to speak.

The removal of one mole of carbon dioxide from exhaled gas generates one mole of water. If all this water vaporizes, and assuming ideal gas conditions exist, an identical volume of water vapour will replace the volume of carbon dioxide. If the exhaled gas contains 5% carbon dioxide, sufficient water vapour could be produced to saturate gas at 33 °C, as saturated gas at 33 °C contains 5% water vapour by volume (36 g m^{-3} water vapour). This is much more than the minimum of 20 g m^{-3} recommended when the upper airways are bypassed during short-term procedures.[8]

Hence, provided low fresh-gas flows are used with the circle breathing system (so that the humidity in the gas is not diluted excessively by the dry fresh-gas), adequate levels of humidity may be produced by the circle system alone (Fig. 11.10). During anaesthesia, the humidity of the gas in the circle system therefore increases, although it may take up to 1 hour or more to reach a maximum level. The humidity in the breathing system will augment the moisture content of the gas delivered to the patient from a filter or HME sited at the patient connection port.[10]

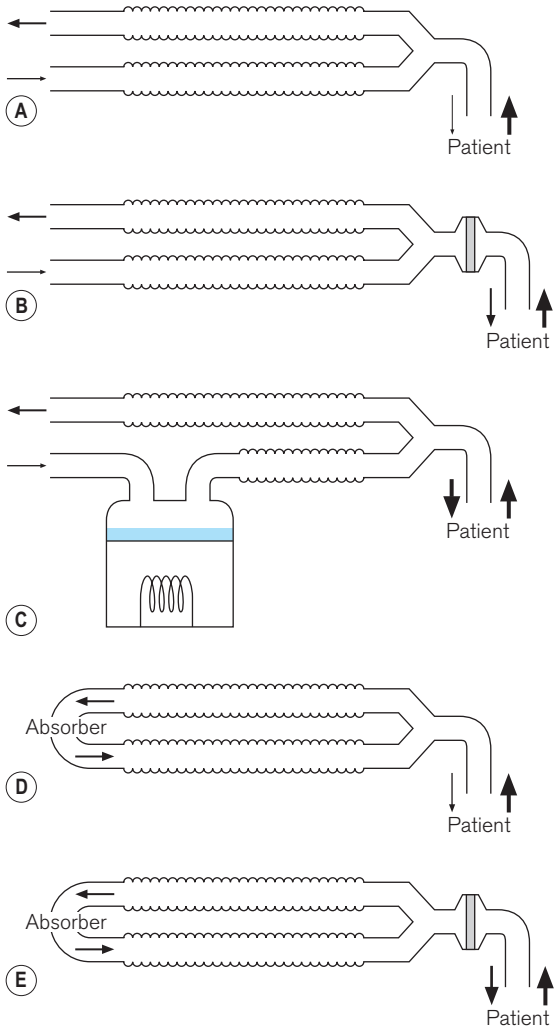


Figure 11.10 Breathing systems with different methods of humidification. **A.** Open breathing system; **B.** breathing system with HME; **C.** breathing system with heated humidifier; **D.** circle breathing system; **E.** circle breathing system with HME. The width of the arrows indicates the level of humidity at various points in the breathing system. High levels of humidity can cause condensation if an appropriate temperature is not maintained.

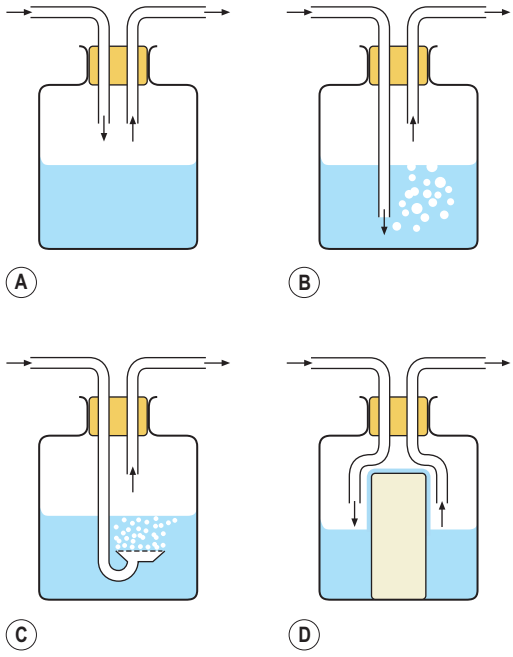


Figure 11.11 Ambient temperature or bottle humidifier. **A.** Simple bottle humidifier; **B.** bubble-through humidifier; **C.** bubble-through humidifier with sintered filter to reduce the size of bubbles and hence increase surface area for evaporation; **D.** wick humidifier.

Active humidification systems

In these devices, water vapour is added to the inspired gasses independently of the patient.

1. Bottle humidifier

The simplest type of active humidifier is the bottle humidifier (Fig. 11.11). In its simplest form, gas is directed over the surface of the water (Fig. 11.11A). The water evaporates and the water vapour mixes with the gas, increasing its humidity. The process is inefficient in that, unless the gas flow is very low, there is not sufficient time for the gas to become saturated with water vapour before it leaves the humidifier. The efficiency of humidification can be improved by increasing the surface area for evaporation. This can be achieved by bubbling the gas through the water, either through a tube (Fig. 11.11B) or through a sintered filter (Fig. 11.11C), or by placing a wick into the water: water is drawn up the wick again increasing the surface area for evaporation (Fig. 11.11D). Bubbling gas through the water increases the pressure drop across the device, and the resistance to gas flow may, therefore, be unacceptably high for a spontaneously breathing patient who is required to draw gas through the humidifier.

However, in all these systems, the ambient temperature limits the maximum level of humidity that can be generated. For typical room air, the maximum level of humidity (humidity at saturation) that can be achieved is about 20 g m^{-3}. However, because the water will cool as evaporation occurs, the level is likely to be less than this. To achieve higher levels of humidity, either the gas or the water (or both) must be heated. Alternatively, a nebulizer may be used (see below).

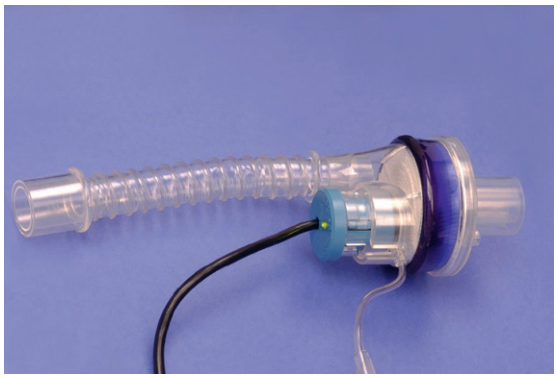


Figure 11.12 The Medisize Hygrovent Gold. Water is gravity-fed into a space between a heater plate and a Goretex membrane. The water is heated and evaporates: the water vapour then passes through the Goretex membrane into the breathing system.

2. Active heat and moisture exchanger

In this device, there is a heat and moisture exchange layer that retains and returns a portion of the exhaled heat and moisture as with passive versions. However, there is also a source of water that is heated within the device so that additional water vapour can be added to the inspiratory gasses. One version of the device can be used with any HME or HMEF, and, therefore, if a filter is used, the patient can be protected from inhaling any infective droplets (Fig. 11.12). This device adds about 5 g m^{-3} to the humidity provided by the HME or filter with which it is used.[11]

3. Heated humidifiers

Heated humidifiers typically consist of the following (Fig. 11.13). Water is heated in a chamber that includes a wick, which absorbs the water and increases the surface area for evaporation. The water vapour produced mixes with the supplied gas as it flows through the humidifier. The water lost by evaporation is either manually or automatically replenished from a reservoir. The humidified gas then flows to the patient through a delivery tube, which contains a sensor that monitors the temperature of the gas at the patient end. The desired temperature of the delivered humidified gas is set using a control knob on the humidifier. If this gas is allowed to cool as it flows through the delivery tube, condensation may occur. To reduce this cooling, a heater wire in the delivery tube maintains the temperature of the gas. The temperature of the gas at the outlet of the humidification chamber is also monitored. The difference in temperature between the humidification chamber and the patient connection port is set on a second control knob. If the heater wire increases the temperature of the gas as it flows through the delivery tube, the risk of condensation forming is reduced, but the relative humidity of the gas also decreases. Alternatively, if the temperature



Figure 11.13 Hudson RCI Conchatherm IV heated humidifier. There are separate controls for 'TEMPERATURE' (temperature of the gas delivered to the patient) and 'TEMP GRADIENT' (difference between the temperature at the patient end of the delivery tube and the humidifier outlet).

of the gas falls, because of the settings on the two knobs, some condensation will occur, but the gas remains fully saturated with water vapour (Fig. 11.14).

Unless a second heater wire is used, condensation will also occur in the expiratory limb. Alternatively, a water trap can be fitted to collect any condensation that forms.

In one type of humidifier, the user has only to select whether the patient is intubated or receiving humidified gasses via a facemask. The temperature and humidity of the gasses are then controlled by the humidifier to provide the optimum level for the patient. The humidifier controls the humidity and temperature and also takes into account the flow of gas (Fig. 11.15).

Nebulizers

Nebulizers are used to humidify respirable gasses by generating aerosols containing droplets of water from a reservoir of water (Fig. 11.16). The water may additionally contain medication. Some of these droplets evaporate as the gas flows to the patient so that the gas is likely to be fully saturated with water vapour when it reaches the patient's airways. However, as heat is required for evaporation, the temperature of the gas will fall. The temperature can be increased by warming the water in the reservoir.

There are two types of nebulizer: gas-driven or ultrasonic. Both types can generate large numbers of water droplets, equivalent to a moisture output of several hundred grams per cubic metre, compared to 44 g m^{-3} for humidity at saturation at 37 °C.

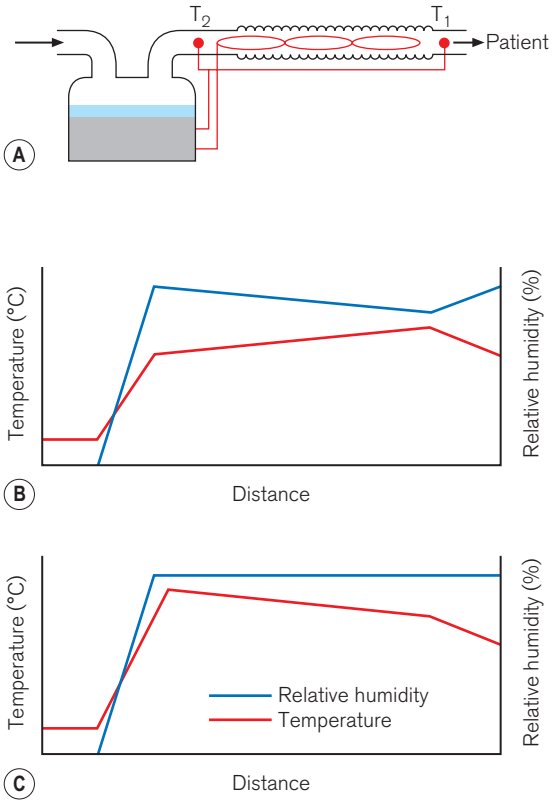


Figure 11.14 **A.** Heated humidifier with heated delivery tube. **B.** Humidifier set so that the temperature of the gas at the patient end of the delivery tube (T1) is higher than the humidification chamber outlet (T2). **C.** Humidifier set so that the temperature of the gas at the patient end of the delivery tube (T1) is lower than the humidification chamber outlet (T2). In this case, condensation will occur in the delivery tube. The humidity of the gas delivered to the patient is the same in both cases.

1. Gas-driven nebulizers

These devices use the Bernoulli principle (see Fig. 7.13) to force driving gas under pressure to entrain water into the system from a reservoir. The water is sucked into the gas stream and is broken up into droplets by the high flow of gas. An anvil, placed in the path of the flow of gas, breaks up the larger droplets into smaller ones, suitable for delivery to the patient (Fig. 11.17A).

2. Ultrasonic nebulizer

In this device, a plate containing a piezo-electric crystal vibrates at an ultrasonic frequency, typically around 2 MHz. Water is either dropped onto the plate, or the plate is placed in the water. The frequency of oscillations of the plate causes the water to break up into droplets. Gas flowing into the nebulizing chamber picks up the droplets to deliver them to the patient (Fig. 11.17B).



Figure 11.15 Fisher and Paykel MR850 heated humidifier. To use the humidifier, the user has merely to choose whether the patient is intubated or receiving humidified gas via a facemask. A heated wire in the delivery tube maintains the temperature of the delivered gas and reduces condensation.

Deposition in the airways

Deposition of droplets in the airways depends on similar mechanisms as deposition of particles in filters (see above). Deposition, therefore, depends on the size of the droplets. Droplets with a diameter greater than 5 µm are deposited in the upper airways (gravitational settling), droplets with a diameter in the range of 2–6 µm are deposited in the tracheobronchial airways (interception), and droplets with a diameter in the range of 0.5–3 µm are deposited in the alveoli (inertial impaction and diffusion) (Fig. 11.18).[12] As with particles and filters, deposition and retention of droplets within the airways is a minimum for droplets with an aerodynamic diameter of about 0.3 µm. The mass of the droplets is also important. The mass of the droplet is proportional to r^3, where r is the radius. A droplet with 10 times the radius will carry 1000 times the mass, or 1000 times the medication. Deposition increases for droplets with diameters less than 0.3 µm due to Brownian motion. However, droplets of this size carry only a very small mass of medication.



Figure 11.16 Aerosol generated by a nebulizer.

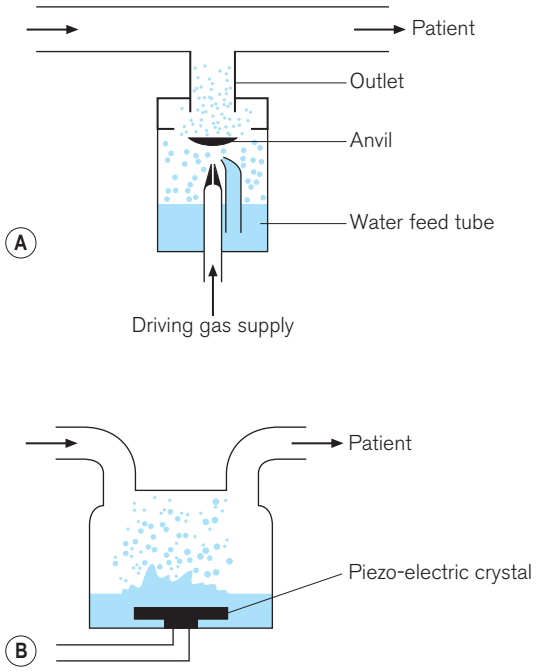


Figure 11.17 Operating principle of the nebulizer. **A.** Gas-powered; **B.** ultrasonic. In **A.** rapid expansion of the gas at the end of the tube causes a reduction in pressure, drawing liquid up the tube, which is then broken up into droplets as it emerges. The droplets are broken into smaller droplets when they strike an anvil. In **B.** a plate vibrates at an ultrasonic frequency (around 2 MHz) which breaks up liquid water into small droplets.

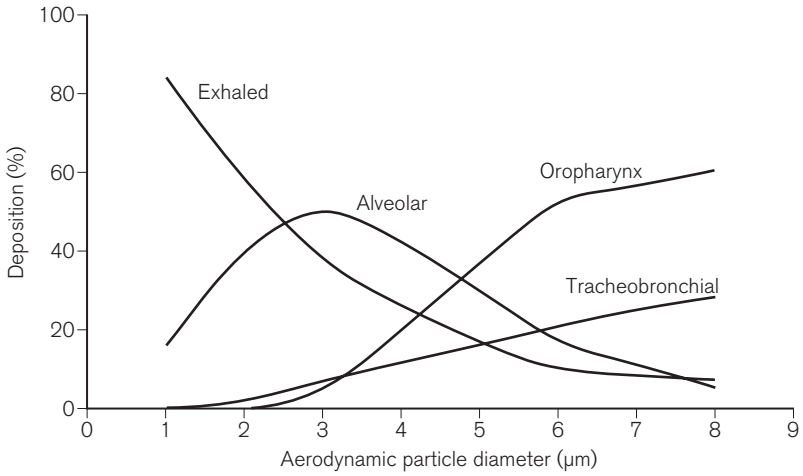


Figure 11.18 Deposition of particles in the airways. Note that maximum deposition occurs in the alveoli for particles with an aerodynamic diameter of approximately 0.5–3 μm. Larger particles are predominantly deposited in the large airways. In this size range, smaller particles are predominantly exhaled.

Data from O'Callaghan C[1] Barry PW. The science of nebulised drug delivery. Thorax *1997;**52**:S25–30.*

Deposition in the alveoli, particularly when breathing nasally, is a small fraction of the overall deposition. Deposition in the alveoli will be greater during oral breathing and particularly when the upper airways are bypassed.

Problems with filters, humidifiers and nebulizers

When added at the Y-piece of a breathing system, filters and HMEs add dead space. A commonly held view is that the HME or filter should add no more than the equivalent of one-fifth of the tidal volume to the dead space, so that for a tidal volume of 0.5 L the maximum dead space of the device should be 100 ml.

Filters and HMEs also increase the resistance to gas flow in a breathing system. Also, if the dead space is reduced, by using a smaller device, the resistance to gas flow increases, and the filtration efficiency and moisture output decreases. Filters can block if excessive water or sputum enter the housing, preventing adequate ventilation of the patient.[13,14]

In some early humidifiers, gas was bubbled through the water in the vaporization chamber. In addition to humidifying the inspired gas, this method produced droplets of water in the gas stream, but which occasionally contained microbes growing in the chamber. This increased the risk of infection to a patient. The risk of infection from modern heated humidifiers is low, as the gas passes over, rather than through, the water.

Heated humidifiers rely on temperature sensors to ensure that the temperature of the delivered gas is that required by the user. The temperature of the gas in the delivery tube is generally warmer in the centre of the stream. If the temperature sensors are not sited correctly in the middle of the gas flow the temperature of the delivered gas may be much greater than that indicated on the humidifier and required by the user and injury may result.

Condensation will occur in the expiratory limb of a humidification system. A heater wire can be used to reduce the condensation. However, this may cause excessive condensation to form within the expiratory valve of the ventilator, adversely affecting the performance of the expiratory flow sensor that may be sited there.

With nebulizers, it is relatively easy to add a large amount of moisture to the delivered gas, leading to excessive loading of the lungs with water and subsequent hypoxia due to blockage of the alveoli. In addition droplets of the sizes produced are very effective carriers of microbes, so care must be taken to ensure that the liquid water in the nebulizer is sterile. Some nebulized drugs can block some types of filter.[15] Also if the nebulizer is operating continuously then a proportion of the medication will also be lost into the expiratory limb of the breathing system during exhalation. This proportion depends on the inspiratory : expiratory ratio of the breathing pattern.

REFERENCES

1. Burton JDK. Effects of dry anaesthetic gasses on the respiratory mucous membrane. *Lancet* 1962;**i**:235–9.
2. Chalon J, Loew DAY, Malebranche J. Effects of dry anesthetic gasses on tracheobronchial ciliated epithelium. *Anesthesiology* 1972;**37**:338–43.
3. Williams R, Rankin N, Smith T, Galler D, Seakins P. Relationship between the humidity and temperature of inspired gas and the function of the airway mucosa. *Crit Care Med* 1996;**24**: 1920–9.
4. Rathgeber J, Keitzmann D, Mergeryant H, Hub R, Zuchner K, Kettler D. Prevention of patient bacterial contamination of anaesthesia-circle-systems. A clinical study of the contamination risk and performance of different heat and moisture exchangers with electret filter (HMEF). *Eur J Anaesthesiol* 1997;**14**:368–73.
5. Wilkes AR. The ability of breathing system filters to prevent liquid contamination of breathing systems: a laboratory study. *Anaesthesia* 2002;**57**:33–9.
6. Wilkes AR, Benbough JE, Speight SE, Harmer H. The bacterial and viral filtration performance of breathing system filters. *Anaesthesia* 2000;**55**:458–65.
7. Wilkes AR. Measuring the filtration performance of breathing system filters using sodium chloride particles. *Anaesthesia* 2002;**57**: 162–8.
8. Kleemann PP. Humidity of anaesthetic gasses with respect to low flow anaesthesia. *Anaesth Intensive Care* 1994;**22**:396–408.
9. British Standards Institution. *Respiratory tract humidifiers for medical use – Particular requirements for respiratory humidification systems (BS EN ISO 8185: 2009)*. London: British Standards Institution; 2009.
10. Henriksson B-Å, Sundling J, Hellman A. The effect of a heat and moisture exchanger on humidity in a low-flow anaesthesia system. *Anaesthesia* 1997;**52**:144–9.
11. Medical Devices Agency. *'Active' heat and moisture exchanger: Tomtec HME-Booster (Evaluation 01029)*. London: Medical Devices Agency; 2001.
12. British Standards Institution. *Respiratory therapy equipment – Part 1: Nebulizing systems and their components (BS EN 13544–1: 2001)*. London: British Standards Institution; 2001.
13. Williams DJ, Stacey MRW. Rapid and complete occlusion of a heat and moisture exchange filter by

pulmonary edema (Clinical report). *Can J Anaesth* 2002;**49**: 126–31.

14. Lawes EG. Hidden hazards and dangers associated with the use of HME/filters in breathing circuits. Their effect on toxic metabolite production, pulse oximetry and airway resistance. *Br J Anaesth* 2003;**91**:249–64.
15. Stacey MRW, Asai T, Wilkes A, Hodzovic I. Obstruction of a breathing system filter. *Can J Anaesth* 1996;**43**:1276.

FURTHER READING

Branson RD, Campbell RS, Davis K, Porembka DT. Anaesthesia circuits, humidity output, and mucociliary structure and function. *Anaesth Intensive Care* 1998;**26**:178–83.

Branson RD, Peterson BD, Carson KD, editors. Humidification: current therapy and controversy. *Respir Care Clin N Am* 1998;**4**:189–344.

Brown RC. *Air filtration. An integrated approach to the theory and applications of fibrous filters*. Oxford: Pergamon Press; 1993.

Hinds WC. *Aerosol technology. Properties, behavior, and measurement of airborne particles*. 2nd ed. New York: John Wiley and Sons; 1999.

Kramer A, Kranabetter R, Rathgeber J, Züchner K, Assadian O, Daeschlein G, et al. Infection prevention during anaesthesia ventilation by the use of breathing system filters (BSF): joint recommendation by German Society of Hospital Hygiene (DGKH) and German Society for Anaesthesiology and Intensive Care (DGAI). *GMS Krankenhaushygiene Interdisziplinär* 2010;**5**:1–19.

MHRA Evaluation 04005. Breathing System Filters. An assessment of 104 breathing system filters. HMSO. London 2004.

Muers MF. Overview of nebuliser treatment. *Thorax* 1997;**52**:S25–30.

O'Callaghan C, Barry PW. The science of nebulised drug delivery. *Thorax* 1997;**52**:S31–44.

Shelly MP. Humidification. In: *Intensive care rounds*. Abingdon: The Medicine Group (Education) Ltd; 1993.

Wilkes AR. Humidification: its importance and delivery. *Br J Anaesth CEPD Rev* 2001;**1**:40–3.

Wilkes AR. Breathing system filters. *Br J Anaesth CEPD Rev* 2002;**2**: 151–54.

Wilkes AR. Heat and moisture exchangers and breathing system filters: their use in anaesthesia and intensive care. Part 1 – history, principles and efficiency. *Anaesthesia* 2011;**66**:31–9.

Wilkes AR. Heat and moisture exchangers and breathing system filters: their use in anaesthesia and intensive care. Part 2 - practical use, including problems, and their use with paediatric patients. *Anaesthesia* 2011;**66**:40–51.

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Chapter 12

Equipment for paediatric anaesthesia

Stephen Fenlon

CHAPTER CONTENTS

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Children neither look nor behave like small adults. Their requirements in the perioperative setting differ, including those of anaesthetic technique and equipment. In the UK, surgery for children constitutes less than 10% of the total surgery performed and, for financial reasons, development of equipment centres on the market for adult patients. Despite this, there is a rich history of innovation in paediatric anaesthesia. Some of the items we take for granted appear too simple to have been the subject of invention, an example of this being the T-piece breathing system developed from Magill's system by Dr Phillip Ayre.[1] It may not be the easiest to use,[2] but lightly modified, remains popular with paediatric anaesthetists the world over (Fig. 12.1).

The differences, both between adults and children and within children of different ages, affect the design of equipment. This is particularly so for those items relating to control of the airway and breathing. Small pieces of equipment designed for use on small patients, must be handled by unwieldy adult hands, and be compatible with international standard fittings. Bulky equipment increases the chance of technical complications, particularly accidental extubation.

ANATOMICAL AND PHYSIOLOGICAL DIFFERENCES BETWEEN ADULTS AND CHILDREN

The magnitude of these differences relate to age. Neonates and infants present the largest variation, the older child increasingly approximates to adult parameters. For the purposes of this chapter, the most important variation is found in the anatomy and physiology of the respiratory system.

Anatomical differences in the airway between adults and children

Compared to the adult, in the child:

- the tongue is relatively large and the larynx is in a higher position
- the epiglottis is longer and U-shaped
- in the younger child, the narrowest part of the upper airway is the cricoid ring: a tube passing easily through the laryngeal inlet may be too tight at the cricoid ring. Should insertion of a tracheal tube be necessary, its fit is critical. Too small a tube leads to an increase in resistance, large leak, and possible fluid ingress around the tube. Too tight a fit creates a risk of mucosal ischaemia and oedema, leading to stridor at extubation
- the larynx is smaller, so the reduction in diameter imposed by a tracheal tube will have a significantly larger effect on airway resistance to flow. The significance of apparatus dead space in comparison to the child's total dead space becomes greater the smaller the child (Fig. 12.2). The two together may accommodate a high proportion of the tidal volume with a significant effect on carbon dioxide elimination.

Physiological differences in breathing between adults and children

- The chest wall of the child is more compliant, and contributes little to ventilation. The diaphragm

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durable plastics are particularly valuable in paediatric breathing systems, providing flexible low-resistance systems with a reduced tendency to 'drag' on other components. For adult patients, single-use breathing systems may be reused, provided an effective airway filter is used to isolate the system and anaesthetic machine from transmissible disease. Evidence is accumulating that paediatric filters are as efficient as the adult versions,[4] but as yet the reuse of breathing systems is not encouraged. For more information see the section on breathing system humidification and filtration later in this chapter.

Regulation of equipment manufacture

The development and testing of new apparatus, and its ease of use, have been reviewed.[5,6] Medical devices sold in the European economic area carry a CE mark (European mark of conformity assessment, Conformité Européene) placed by the manufacturer. To achieve this, the manufacturer provides details of risk analyses, performance in standard tests and technical data relating to manufacture of an item of equipment. The Competent Authority, for the UK, the Medicines and Healthcare Products Regulatory Agency, oversees this procedure (see Chapter 28). Medical devices are classified and tested according to potential risk of injury, e.g. a facemask is class 1 (low risk), a cardiac catheter class 3 (high risk).[7] The CE marking process does not imply specific clinical testing; most pre-use testing is so-called bench testing, demonstrating equivalence or better function than existing similar equipment. Under these rules, scaling down of adult equipment to paediatric size is acceptable, but may not produce the most effective devices in use. An urge to release a new device meeting minimum standards onto the market is balanced against the need for commercial success; this provides manufacturers with an incentive to produce equipment with demonstrable clinical value. As an example, the laryngeal mask whilst scaled down from adult versions was still subject to specific testing to confirm it retained anatomical suitability for paediatric use.[8] Further versions of this device have been subject to post marketing tests of performance in the clinical environment.[9,10] A small number of devices without a CE mark are used on patients; these may be custom built or those requiring more clinical testing before a CE mark can be authorized. A device made within a hospital, for use in that hospital, does not require a CE mark, and may be provided with an exemption certificate for specified use elsewhere. In summary, excepting a small number of unique devices, all medical equipment used on children in the UK has a CE mark. This mark provides reassurance of manufacturing standards but does not imply clinical effectiveness, for which independent evaluation should be sought.

Equipment for management of the airway

Apparatus for management of the paediatric airway, from facemasks through to tracheostomy tubes, is outwardly similar to the adult equivalent. Management of the airway in both adult and paediatric practice has been revolutionized by the introduction of the laryngeal mask airway. Similar airway management devices introduced following the laryngeal mask have not so far enjoyed the same level of success.

Facemasks

These should be available in a range of appropriate sizes and form a good seal at the edges, with minimal dead space. Clear plastic masks are less frightening to awake children and can even be scented, though matching the overpowering bouquet of volatile anaesthetic agents presents a major challenge for any perfumier.

A variety of paediatric facemasks exist (Fig. 12.3). To reduce dead space, the Rendell-Baker-Soucek mask was designed anatomically, from casts of children's faces in the same way as a dental plate is made.[11] This mask achieves a seal by virtue of its close approximation to the contours of the face. Other masks require some form of flexible lip or air filled cushion. The lipped round silicone mask (Fig. 12.3E) is easy to apply, providing an excellent seal for infant use. Disposable masks generally employ a cushion seal, the rest of the mask being of rigid construction. Whichever is chosen, it must be easy to hold and seal on the face, and this may well be a matter of trial and error. Attempts to reduce facemask anatomical dead space may be less important than previously thought, the actual

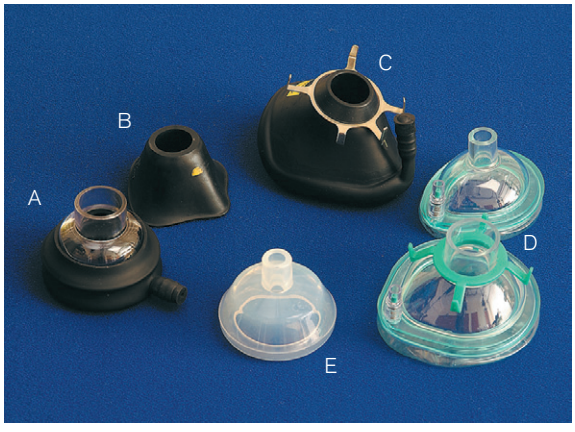


Figure 12.3 A gathering of facemasks. **A.** The Ambu facemask. **B.** The Rendell–Baker–Soucek facemask. **C.** The anatomical facemask. **D.** Disposable scented facemasks. **E.** The Laerdal silicone facemask.

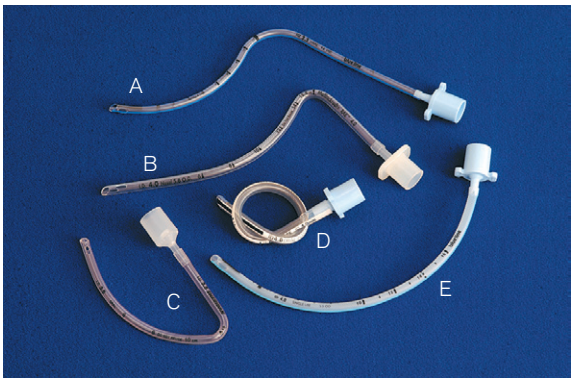


Figure 12.4 An assembly of tracheal tubes. **A.** Oral north facing tube; **B.** RAE tube; **C.** RAE south facing oral tube; **D.** reinforced tube; **E.** standard tracheal tube.

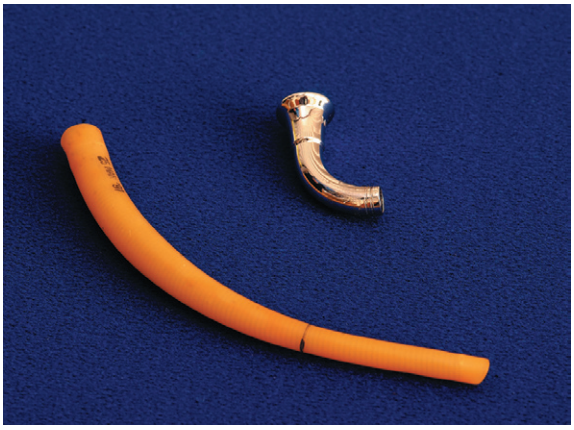


Figure 12.5 The Enderby tracheal tube, a tapered tube designed for use in cleft lip and palate surgery (no longer in use).

increase in physiological dead space with anaesthesia being less than predicted.[12]

Tracheal tubes

Tracheal tubes are available in sizes and shapes to suit different patients and surgical procedures (Fig. 12.4). The internal diameter of the tracheal tube is the major determinant in airway resistance and hence the size by which tubes are measured and selected. The fit of the tube to each patient is determined by external diameter which varies with tube wall thickness. This is itself determined by the type of tube (Table 12.1), but can also vary for the same tube from different manufacturers.[13] Other factors affect tube resistance: connectors, tube length, shape of tube and tendency to collect secretions.

The decision to intubate, and which tracheal tube to use, is of great significance. Previous attempts to circumvent the problem of tube resistance included the use of shouldered and tapered tubes, but these designs have been shown to confer no real advantage (Fig. 12.5). Below the age of 10 years, uncuffed tracheal tubes were the norm and were believed to minimize the chance of mucosal damage and post extubation stridor. Despite this perceived advantage, the lack of an airway seal with uncuffed tubes can permit fluid to enter the tracheobronchial tree, contribute to atmospheric pollution, lead to inadequate ventilation and induce anaesthesia in surgeons working around the upper airway.

Widespread use of uncuffed tubes has been questioned.[14] Cuffed tubes offer advantages (Table 12.2) and are safe when used appropriately.[15] New and better designs of cuffed paediatric sized tubes are emerging but more work still remains to be done, particularly on cuff position in preformed (shaped, e.g. RAE type) tubes and on the relationships between tube length and diameter.[15] A change in practice will take time, and selecting an uncuffed tube with a leak at an inflation pressure of 25 cm H_2O will remain common practice, despite evidence that in short procedures at least, it confers no benefit.[16]

Tube size selection is critical particularly for uncuffed tubes; formulae provide only a guide to the correct tube size (Table 12.3). Coexisting medical conditions may influence tube size, for example: children with Down syndrome often require a tube 1–2 mm smaller than expected for their age.[17] Likewise, the required length of tube can only be estimated. Some tubes incorporate marks intended to guide how far to advance the tube into the larynx under direct vision. Preformed tubes may have a mark indicating the position for fixation over the lip. The placing of such marks is inconsistent across tube sizes and manufacturers, and they should not be relied upon.

Fixation of the tube should aim to prevent displacement, maintain the tube position with head movement, and still be relatively easy to secure and adjust (Fig. 12.6). Simple tape fixation fulfills many of these criteria (Fig. 12.7). Nasal intubation in children is more secure and preferred in the intensive care setting, as the tube tends to move less, reducing trauma to the tracheal mucosa.

Flow at the interface of breathing system and tube is disturbed by changes in diameter and direction. Connectors aim to minimize this by smooth internal surfaces, gradual reductions in diameter and gentle direction changes. The commonest tube connector is the ISO 15 mm; another ISO standard system based on 8.5 mm connectors (Fig. 12.8) appears rarely used. Connectors do not reduce the available lumen as they dilate the tube at the point of insertion. Problems can arise when assembling small thin-walled parts, with buckling of the walls and possible occlusion of the lumen.[18] Some older connectors remain in use as they are compact and may offer less resistance to gas flow (Fig. 12.9). Endotracheal tubes allow suction to be applied to the lower airway. To size a suction catheter for use, doubling the tube diameter in mm, gives the appropriate French gauge catheter size.[19]

Table 12.1 Dimensions of some non-cuffed infant endotracheal tubes

MANUFACTURER	INT. DIAMETER (mm)	EXT. DIAMETER (mm)
Portex (silicone)	2.5	3.4
Sheridan (Ped-soft)	2.5	3.6
Portex (ivory)	2.5	3.6
Mallinkrodt (PVC)	2.5	3.6
Rusch (clearway)	2.5	4.0
Mallinkrodt (reinforced)	2.5	4.0
Portex (reinforced)	2.5	4.0
Rusch (rubber)	2.5	4.0
Portex (silicone)	2.5	4.2
Sheridan (Ped-soft)	3.0	4.2
Portex (ivory)	3.0	4.4
Mallinkrodt (PVC)	3.0	4.3
Rusch (clearway)	3.0	4.7
Mallinkrodt (reinforced)	3.0	4.7
Portex (reinforced)	3.0	4.7
Rusch (rubber)	3.0	4.7
Portex (silicone)	3.5	4.8
Sheridan (Ped-soft)	3.5	4.9
Portex (ivory)	3.5	5.0
Mallinkrodt (PVC)	3.5	4.9
Rusch (clearway)	3.5	5.3
Mallinkrodt (reinforced)	3.5	5.3
Portex (reinforced)	3.5	5.3
Rusch (rubber)	3.5	5.3

Table 12.2 A comparison of cuffed and uncuffed tubes

	UNCUFFED TRACHEAL TUBE	CUFFED TRACHEAL TUBE
Seal	attempts a seal in the cricoid ring, but seal may not be effective	used appropriately forms a good seal below cricoid in larger diameter of trachea
Effect on airway mucosa	may be less prone to causing damage	less evidence of long term safety currently available
Available lumen	maximizes available lumen of artificial airway	may reduce the available lumen
Other		- allows use of smaller diameter tubes which may impact post-intubation laryngeal morbidity - choice of tube diameter less critical - lower tube exchange rates - new designs still needed with tube dimensions revised for new paradigm of cuffed tubes.

Table 12.3 A guide to tracheal tube dimensions

AGE	WEIGHT (kg)	TUBE DIAMETER (mm)
Preterm	<2	2.5
Small term	2–3	3.0
Term	3–3.5	3.5
3 months to 1 year	5–10	4.0
1 to 2 years	10–15	4.5
Over 2 years according to formula:		(Age/4) + 4.5 mm
Tube length (oral)		(Age/2) + 12 cm
Tube length (nasal)		(Age/2) + 15 cm

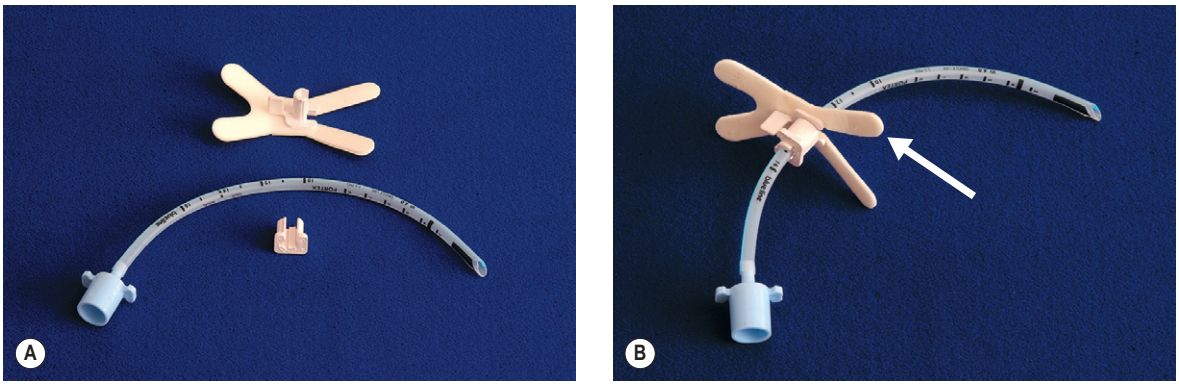


Figure 12.6 **A.** A device for fixing tracheal tubes, The Portex RSP **B.** The same device locked onto the tube, self-adhesive strips (arrowed) are used to fix the device to the face.

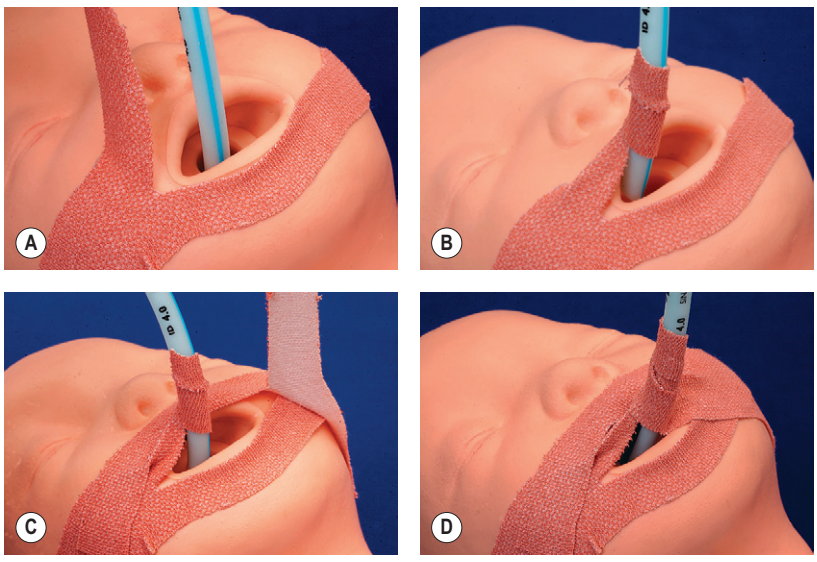


Figure 12.7 Simple tube fixation with adhesive tape.

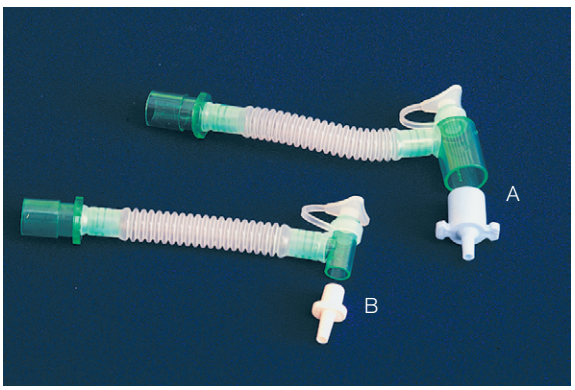


Figure 12.8 ISO fitting tracheal tube connectors and catheter mounts. **A.** 15 mm; **B.** 8.5 mm. Both are available for a range of paediatric tube sizes.

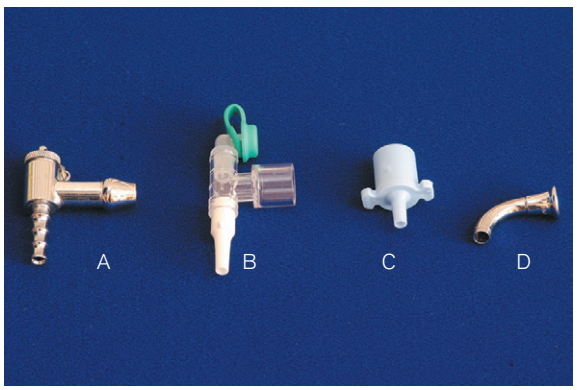


Figure 12.9 Tracheal tube connectors. **A.** Oxford; **B.** ISO 8.5 mm; **C.** ISO 15 mm; **D.** Magill oral.



Figure 12.12 Various sizes of paediatric oropharyngeal airway, in order of increasing size: 000, 00, 0, 1.

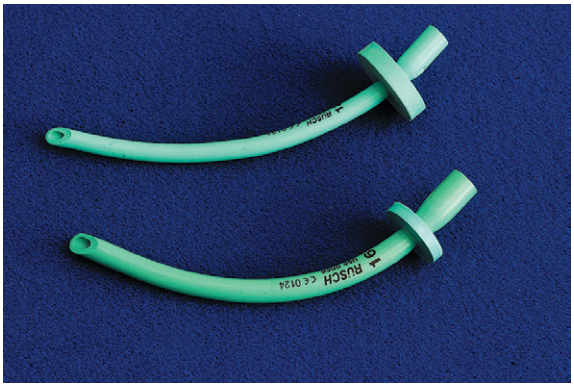


Figure 12.13 Paediatric adjustable flange nasopharyngeal airways.

procedure, compared with jet entrainment or apnoeic techniques. Care is needed with laryngeal mask cuff pressures, particularly if nitrous oxide is employed; unchecked pressures are usually higher than expected and may injure structures in and around the upper airway.[30] Overall, the complication rate is low and the laryngeal mask and its variants have a place in anaesthesia for an increasingly wide range of paediatric patients.

Other supraglottic airway devices

A number of other supraglottic airways have followed in the wake of the laryngeal mask. Their acceptance has been gradual and unbiased evaluation scarce. The outcome of appeals for new initiatives in further independent assessment of clinical performance is awaited.[31]

Airway adjuncts

Scaled versions of oral and nasal airway adjuncts exist for paediatric use (Figs 12.12 and 12.13) and are discussed in Chapter 6.



Figure 12.14 An uncuffed tracheostomy tube size 4 mm internal diameter.

Tracheostomy tubes

A full range of uncuffed tracheostomy tubes exists for use in children (Fig. 12.14). To avoid endobronchial intubation, the intratracheal length is kept short; hence accidental decannulation is easily achieved.

Gaining access to the airway

Airway instrumentation and visualization differs in paediatric practice due to the anatomical differences previously mentioned. Difficulty visualizing the airway is unusual in paediatric practice overall. Management plans need to take account of the additional challenges posed by small and rightfully uncooperative patients.

The laryngoscope

The larynx is usually seen with the direct laryngoscope. A variety of laryngoscope blade profiles exist. The choice is usually dependent upon the age of the patient and the personal preference of the anaesthetist (Fig. 12.15). Many practitioners use a straight blade for infant laryngoscopy; this blade picks up the relatively large epiglottis, affording a better view. For older children, a curved (small Macintosh pattern) blade will suffice. The little finger of the hand holding the laryngoscope may be used to apply external laryngeal pressure to improve the view. Flexible tracheal tube introducers can be used to railroad a tube into a larynx when a direct view cannot be obtained. These are available for use with tubes as small as 2.5 mm internal diameter.

A development of the bougie is the airway exchange catheter (AEC) (Fig. 12.16). A semi-rigid, hollow catheter with interchangeable 15 mm ISO male and Luer-Lok connectors, this is designed to allow tracheal tubes to be exchanged whilst retaining the ability to oxygenate the lungs. The AEC is inserted through the tracheal tube, which can then be removed and exchanged over this. Once in

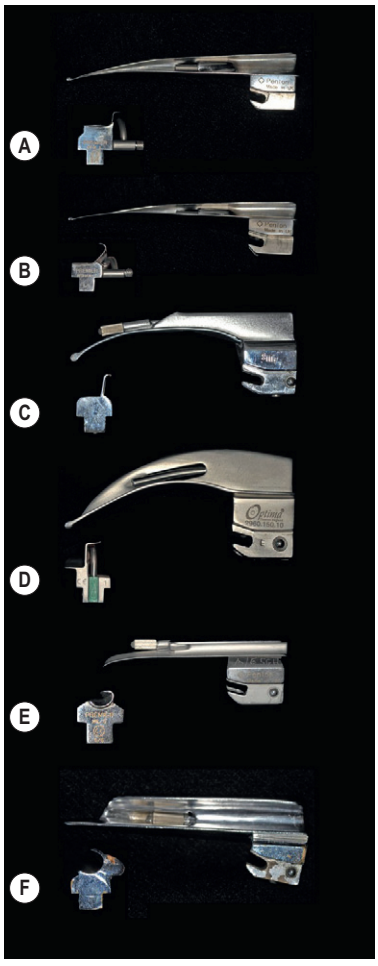


Figure 12.15 Paediatric laryngoscope blades in profile and end-on views. **A.** Seward, **B.** Robertshaw, **C, D.** different interpretations of Macintosh profile, **E.** Miller, **F.** Wisconsin. There are no standards defining the profile of each design, hence variations between manufacturers are to be expected.

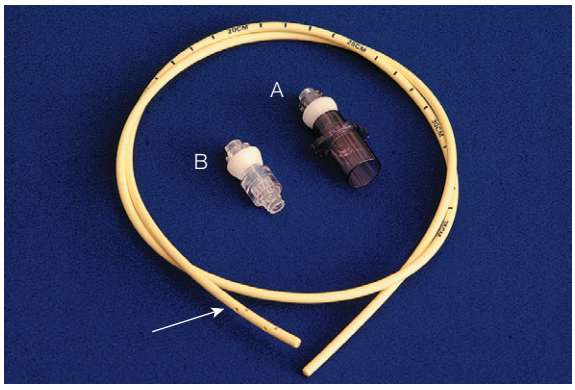


Figure 12.16 An Airway Exchange Catheter (Cook Medical) suitable for paediatric use. The distal end with side holes is marked with an arrow. Also shown are the proximal end Rapi-Fit connectors, **A.** ISO 15 mm; **B.** Luer-Lok.

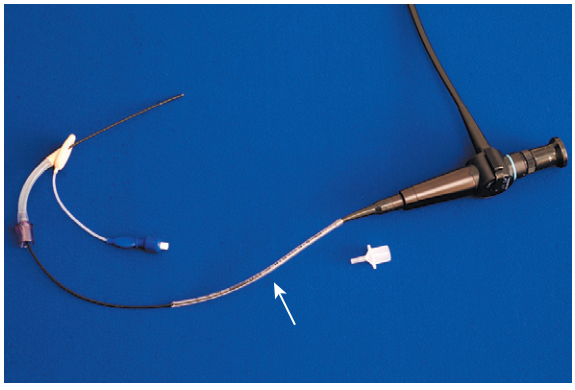


Figure 12.17 The tracheal tube (arrowed) with connector removed, is loaded onto the bronchoscope, which is then advanced into the trachea using a laryngeal mask as guide.

place, the end connector can be fastened, gas sampled from the tip to detect carbon dioxide and correct position confirmed. Oxygen can then be injected or insufflated through the catheter with either a high-pressure injector or a standard anaesthetic breathing system. With the connector removed, a tracheal tube can be guided over the catheter and into the trachea.[32] The catheters are available to fit within tracheal tubes down to 3 mm internal diameter.

The fibreoptic bronchoscope

Both the standard size and a smaller 2.2 mm, flexible fibreoptic bronchoscope can be used to aid intubation in children when the larynx is difficult to visualize. The standard bronchoscope incorporates a suction channel, which can be used to thread a guide wire that is advanced into the trachea under direct vision. After removal of the bronchoscope, the wire remains in the trachea and is used to railroad a tracheal tube of the appropriate size.[33]

Use of the 2.2 mm bronchoscope follows the adult pattern of fibreoptic intubation, whereby the tracheal tube (down to size 2.5 mm) is 'loaded' onto the bronchoscope prior to visualizing the laryngeal inlet. This technique is suited to difficult intubation in infants. A tracheal tube without its connector is passed along the bronchoscope (Fig. 12.17). Following induction of anaesthesia, the airway is maintained with a laryngeal mask. The right-angle connector in the breathing system incorporates a sealing port, through which the bronchoscope is advanced, passing through the bars of the laryngeal mask and into the trachea. The cuff of the laryngeal mask is deflated and withdrawn back along the bronchoscope. The tube is

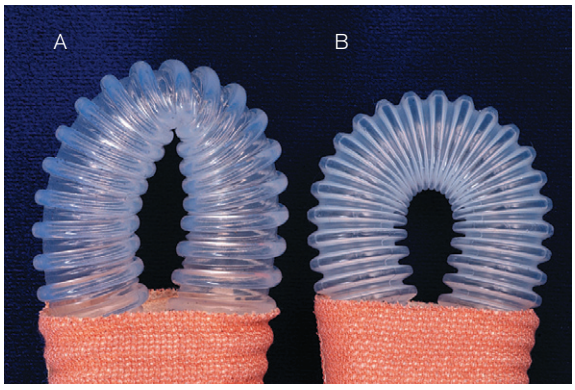


Figure 12.20 Deformation and reduction of tube lumen on extreme flexion. **A.** Coil-over smooth bore tube; **B.** standard corrugated tube.

dead space by a particular piece of equipment may differ from its measured volume, due to the effects of mixing by gas flows within the item. The term functional dead space incorporates this phenomenon.

Partial rebreathing of exhaled gas is a feature of some breathing systems, and is described in more detail in Chapter 5. The T-piece, classified as Mapleson E system, is an example of this. It is worth noting that it is similar in function to the D system.

Resistance is reflected in the pressure gradient required to drive gas through the breathing system to and from the patient. This translates to work done by the patient when breathing spontaneously. High resistance increases work and is tolerated particularly poorly by infants. Resistance arises from the components through which gas flows, including valves where fitted. As with apparatus dead space and rebreathing, resistance should be minimized where possible. This is best achieved by use of a valveless breathing system (such as a T-piece), avoiding acute angulation of connectors, and careful choice of tracheal tube (the source of greatest resistance). Where valved breathing systems are in use, the valves should be designed to be as light as possible.

Repeatedly reversing the direction of flow of relatively large volumes of gas provides additional inertial resistance in the system, which is reduced with by use of smaller-bore hoses. The latter may have a reinforcing coil placed around the outside (in place of the standard corrugated type) permitting a smooth internal surface, with lower resistance to gas flow. Although all tubing is designed to show marked resistance to deformation, this form may kink more easily on extreme flexion (Fig. 12.20).

Compression volume (Table 12.5) during positive pressure ventilation refers to a combination of breathing system distension and internal compression of gas within that system caused by a pressure rise during inspiration. This will affect the inspiratory volume delivered to the patient (see below).

Table 12.5 Compression volume

Definition	Volume lost during positive pressure ventilation due to compression of gas within the system
Features in effects	All breathing systems More significant with large volume systems More significant with small tidal volumes Can be compensated for

During a positive pressure breath, the pressure in the breathing system is raised above atmospheric. This has two effects. Components such as the hoses of the breathing system tend to distend, but, more importantly, according to Boyle's law, the gas within the system will reduce in volume. This is more noticeable, the greater the volume of the system. The overall result is that some of the gas intended for the patient remains in the breathing system and a smaller tidal volume is delivered at the patient end than that introduced into the system by the ventilator. The effects of compression volume are seen with all patients when the lungs are ventilated with positive pressure, but its significance is greater for children due to their smaller tidal volume in relation to total system volume.

As an example, consider an older child attached to a circle system of 5000 ml total internal volume. The ventilator adds 500 ml of gas to the system for inspiration, causing a pressure rise of 20 cm H_2O in the system. Taking atmospheric pressure to be 1000 cm H_2O, this pressure increase represents a fractional rise of 20/1000 cm H_2O or 2%. As pressure increases, volume decreases by the same proportion; here 2% of 5000 ml is 100 ml. Of the 500 ml added by the ventilator, 100 ml has been 'lost' to the compression volume, together with a further smaller loss to system compliance, leading to just under 400 ml reaching the patient. In this example, 20% of the inspiratory volume added by the ventilator is lost to compression.

Now consider a smaller child requiring a tidal volume of 100 ml. Assuming the same pressure increase occurs on inspiration, the same volume is lost to compression and a total volume of 200 ml would need to be added to the system to achieve a tidal volume of 100 ml. Now, 50% of the inflation volume is lost to compression, thus for smaller patients a proportionately larger volume is added to the system in inspiration to maintain adequate tidal volumes.

Increased airway resistance or reduced lung compliance will increase system pressure in a patient ventilated with controlled volumes. At higher pressures, more volume is lost to compression, and the volume delivered to the patient falls, sometimes termed *'preferentially ventilating the compression volume'*.

Table 12.7 Advantages of the circle system

Very low fresh gas consumption
Reduces pollution
Warms and humidifies gasses

In current use, the perceived advantages of circle systems are a reduction in fresh gas flow, easy scavenging, reduced atmospheric pollution and addition and conservation of some heat and water vapour to inspired gas. When changes from adult to paediatric set-up are required, using the same system reduces the chances of assembly errors. However, if spontaneous ventilation is to be used in the smaller child it is perhaps worthwhile changing the breathing hoses from 22 mm to 15 mm diameter, and using smaller distal connectors with minimal dead space.

The circle system has been used successfully in children and infants, the latter usually with controlled ventilation.[46] Reduction in fresh gas flow is less significant in small children, where all breathing systems employ a relatively low flow anyway.

Additionally, leaks from the system around the tracheal tube or laryngeal mask will limit flow reduction, though in practice this rarely precludes efficient use of the circle system.[26] Gauging and rapidly changing the concentration of oxygen and volatile agent in the system is less easy than with the T-piece, and monitoring of these variables is mandatory. The large compression volume is reduced to some extent by the use of 15 mm tubing, but remains greater for this system.

The circle is a more complicated system than others used in paediatrics. However, used with appropriate monitoring and vigilance, it provides the added benefit of, amongst others (Table 12.7), significant reduction in wastage of anaesthetic agents.[47]

Breathing system humidification and filtration

Humidification of dry inspired gas by the patient is an important source of heat loss. Dry inspiratory gasses increase tracheal mucosal damage and the incidence of tracheal tube blockage. A range of simple heat and moisture exchangers is now available and in common use by anaesthetists (see Chapter 11). The level of humidification and heat retention necessary to prevent the above problems is unknown. Furthermore, the performance of humidifiers varies significantly between manufacturers and during use.[48]

Anaesthesia circuits may become contaminated with microbes from the patient's respiratory tract; hepatitis C virus transmission has been reported between patients[49] linked to breathing system contamination. Combined airway humidifiers and filters aim to prevent microbes entering the breathing system, whilst conserving airway heat and humidity (Fig. 12.23). Their widespread adoption into clinical practice may be beneficial, but should receive some appraisal. Filters contribute dead space and resistance to the breathing system, and may lead to more serious complications.[50] These effects are greater for smaller tidal volumes, young children in particular require careful observation.[51,52] In addition to such problems when using filters in children, there are doubts about the efficiency of these products. On standardized assessment, filters from various manufacturers exhibit differing particle penetration.[53] This may result from a failure to account for the lower flow rates pertaining in paediatric systems,[4] but in practice means paediatric airway filters have not been shown to offer system isolation to the same extent as their adult counterparts. Additionally it would seem that even very modest hydrostatic pressures of 0.3–0.5 kPa can cause liquid penetration of filters, which may carry with it infective agents.[54] A recent study was able to demonstrate bacterial contamination of the machine (breathing system) side of filters in 9% of sampled filters.[55] At the time of writing, knowledge of airway filtration is insufficient to recommend that airway filters allow reuse of breathing systems in children.

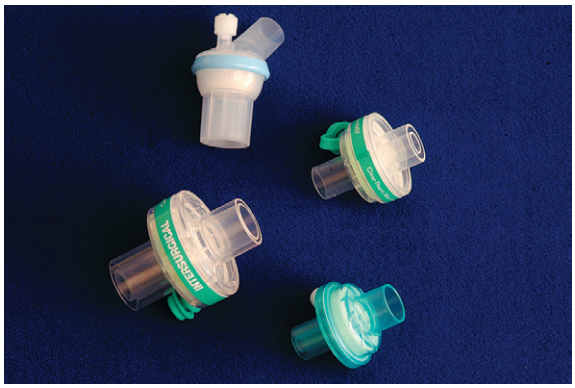


Figure 12.23 A selection of humidifiers and filter/humidifiers for paediatric use.

Ventilators for use during paediatric anaesthesia

Ventilators are described and classified in more detail in Chapter 9.

Ventilators designed specifically for paediatric use should ideally have a low internal volume and resistance, the ability to deliver small tidal volumes and high respiratory rates and a pressure-control feature. However, increasing sophistication of microprocessor control has allowed ventilators with larger internal volumes, e.g. the Penlon AV900 (Fig. 12.24), to be suitable for both paediatric and adult patients and most modern anaesthesia workstation ventilators are currently suitable for use with small children.

Table 12.9 Advantages of volume control

Set tidal volume delivered
Tidal volume maintained with changes in resistance or compliance
Pressure dependent alarms activated with above changes

latter should have a minimum volume of 350 ml to prevent the driving gas from diluting the inspired gas.

In the inspiratory phase, gas from the ventilator enters the body of the valve according to the set parameters for flow and time. From here it can pass two ways, through the 3.5 mm diameter orifice to atmosphere, or into the patient limb of the valve. By altering the flow of gas from the ventilator, the pressure within the valve is controlled. The result of increasing the flow of driving gas is described below:

- At the recommended fresh gas flow rate for the T-piece and at a low inspiratory flow rate from the ventilator, the pressure developed inside the Newton valve is low as a result of the continuous leakage from the fixed orifice outlet. Therefore, the valve only partially dams the outlet of the breathing system and so acts as a 'partial thumb occluder'. This transmits a small tidal volume to the patient at a rate depending on, but less than the fresh gas flow into the T-piece.
- As the flow of driving gas from the ventilator increases, at some stage, inflow to the valve will be equal to the leak through the orifice and at this time the valve behaves almost as a complete thumb occluder on the T-piece expiratory limb. The delivered tidal volume then equals the fresh gas flow to the T-piece.
- Further increasing flow from the ventilator results in flow into the valve exceeding the leak, and some of the driving gas now passes back along the T-piece expiratory limb and can act as a gas piston. Tidal volumes will now exceed fresh gas flow and are altered by ventilator settings.
- During the expiratory phase, all gas passes out via the valve orifice, and for expiratory flows up to 15 l min^{-1} pressure within the valve should not rise above 5 cm H_2O.

Modified with a Newton valve, the Nuffield 200 Series ventilator may be seen as the mechanical equivalent of the anaesthetist's hand in combination with the open-ended bag on a T-piece. The system is easy to understand, can be switched rapidly from manual to automatic ventilation and permits scavenging of waste gas. It can deliver tidal volumes between 10 and 300 ml at frequencies from 10 to 85l min^{-1}, making this a suitable ventilator for neonates and infants. The system requires high fresh gas flows and consumes large amounts of pressurized gas to drive the



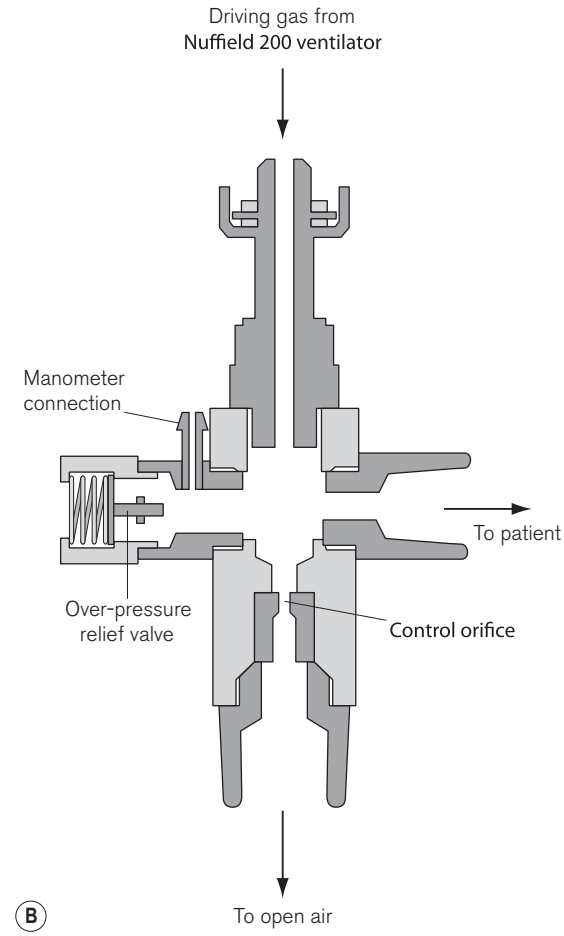


Figure 12.25 **A.** The Penlon Nuffield Series 200 ventilator. Shown inset, the fixed expiratory orifice of the Newton valve. **B.** Sectional view.

ventilator. The Newton valve is not suitable for patients over 20 kg.

Access to the circulatory system

Obtaining and maintaining secure venous access in small patients can be very challenging. As with adults, there are essentially two means to achieve this: the cannula over needle or wire through needle (Seldinger) techniques. Use of rigid indwelling needles alone is not recommended, as they tend to cut out of the vein resulting in extravasation of administered fluids. Much smaller cannulae are required for both peripheral and central access. Flow rates through narrow-bore cannulae are low (Table 12.10). In an emergency it may be impossible to cannulate a vein. Below the age of 6 years an intraosseous needle (Fig. 12.26) provides alternative access to the circulatory system for administration of drugs and fluid until venous cannulation is achieved. All contaminated sharps must be disposed of safely, and some cannulae incorporate a retraction device to ensure the needle end is covered after the vein is entered (Fig. 12.27). Cannulation of central veins, particularly the internal jugular and femoral veins, has traditionally been undertaken using surface landmarks as a guide. The use of two-dimensional ultrasound to cannulate central veins is well described.[56] Ultrasound has been suggested to improve the success and safety compared to landmark techniques.[57] Where long-term venous access is required, peripherally inserted lines may be easier to site and can be threaded into central veins if necessary.

Positioning, environmental control and temperature monitoring

Great care is required with positioning during anaesthesia, including eye protection and protection of vulnerable peripheral nerves, etc. Padding under the shoulders of the supine infant prevents the large occiput from putting the head into flexion. Depending on the site of surgery, a bar or bridge placed over the patient prevents the surgeon from inadvertently leaning on the chest or face and allows improved access for the anaesthetist to monitor the patient, check line sites, etc. When a limb tourniquet is used, they must be of adequate width and exceed limb circumference by 7–15 cm. Padding is needed, particularly at the edges, and the tourniquet can be inflated to a lower pressure than that for adults. Skin preparing fluids must not soak under the tourniquet, as the child's thinner skin is easily damaged.[58]

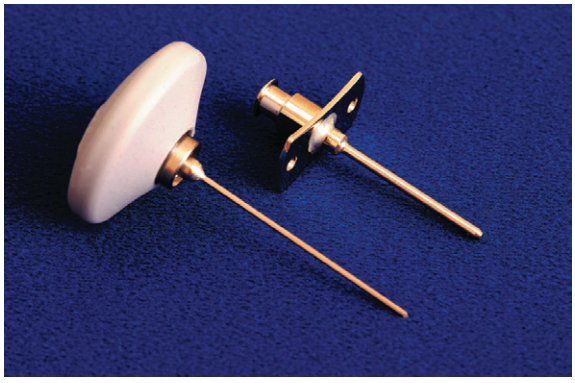


Figure 12.26 An intraosseus needle. On some designs the metal cannula has an outer screw thread for fixation into the cortical bone.

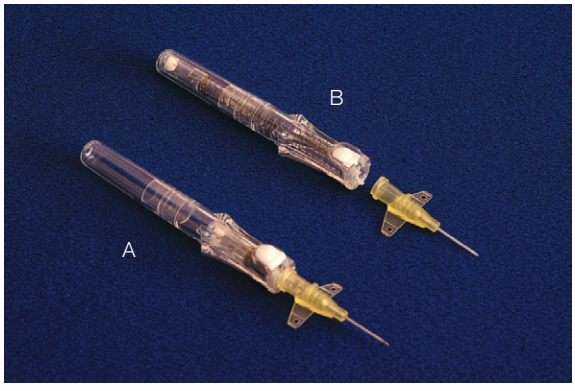


Figure 12.27 A sharp safe needle, the Insyte–N Autoguard. **A.** The cannula and needle prior to insertion; **B**. following successful venepuncture, the needle is retracted into the clear hub.

Table 12.10 Flow rates (ml min^{-1}) through intravenous cannulae

CATHETER SIZE (SWG)	CRYSTALLOID (GRAVITY)	CRYSTALLOID (PRESSURE)	BLOOD (PRESSURE)
24-gauge	14–15	42–47	20–30
22-gauge	24–26	65–77	44–50
20-gauge	38–42	103–126	69–81
18-gauge	55–62	164–214	150–164



Figure 12.28 Foil wrap (Mediwrap).



Figure 12.29 A child manikin lies still atop a forced air warming mattress (Bair Hugger). Clear plastic drapes help to create a warm microclimate. The hole at the top of the mattress (arrowed) connects to the warm air source.

All patients can lose heat during anaesthesia, the thermoneutral temperature zone (about 28 °C in an unclothed adult) being higher in neonates. Small children have limited thermogenesis, so heat loss may be difficult to recoup.[59] Depending on the operation and exposure, heat is lost through a combination of radiation, convection, conduction and evaporation of bodily fluids and can be prevented by a number of measures:

- Maintain the operating theatre at a higher temperature and humidity; this is rarely within the limits of staff comfort.
- Reduce evaporative loss by limiting exposure of wet areas at the operative site and humidifying inspired gas.
- Reduce radiation loss with foil blankets; correctly applied these reduce convective loss too (Fig. 12.28).
- Active warming, most commonly in the form of forced air warming, is a more effective way of maintaining temperature.[60] A warm air microclimate can be created around the child by use of forced warmed air, in combination with impervious clear plastic covers adapted for surgical access (Fig. 12.29).

Whatever methods are chosen, reliable monitoring is required, as overheating of small children is easily achieved, and should be considered mandatory if active warming methods are used even for a short time. Oesophageal, rectal, axillary and tympanic membrane temperatures all correlate well to central temperature. For operations where a urinary catheter is needed, this can incorporate a temperature probe at the tip, providing an excellent means of monitoring.

Transfer of the critically ill child

An increasing trend for transferring critically ill children to centralized paediatric intensive care units requires safe and rapid transfer conditions, usually provided by a retrieval team from the accepting unit. Resuscitation is initiated at the referring unit, often under the guidance of the retrieval service. A challenge for practitioners with little exposure to sick children is to judge appropriate tube sizes, etc. This is made easier by some form of universal measuring device together with prepared packs of equipment,[61] such as the Broselow Tape and the Broselow Paediatric ALS Organizer.

Transfer brings risks from moving a potentially unstable patient into an environment compromised by motion, cold, poor light, noise and limited electric power. Ideal equipment for this is lightweight, robust and compact, and has a reliable power supply. All equipment can fail and manual back-up, such as self-inflating bags, is vital. Transfer equipment is kept together in a series of clearly identified portable packs (Fig. 12.30); all batteries are continuously charged and packs are replenished and checked for immediate reuse after a transfer. Equipment requirements vary according to the size of child and the nature of the illness, but it is likely that ventilation and sedation will be needed together with inotrope infusions and intravenous fluids.

Portable ventilators (Fig. 12.31) need to be capable of delivering air/oxygen mixes, creating positive end expiratory pressure, displaying airway pressures, detecting and alarming for disconnections and must have adjustable tidal volumes for a range of patient sizes. They are usually driven from a high-pressure oxygen source and have high gas consumption (20 L min^{-1} for the example shown). Tubes and lines need to be secured sufficiently well to survive multiple transfers and movement. Temperature control can be difficult; rigorous use of foil helps, and highly insulating material such as 'bubble wrap' provides excellent thermal insulation. Single-use heated mattresses are available for infants, heated by initiating a chemical reaction of the contents. Compact single module monitoring is available and even blood biochemistry

REFERENCES

1. Ayre P. Endotracheal anaesthesia for babies; with special reference to hare-lip and cleft palate operations. *Current Research in Anesthesiology* 1937;**16**:330.
2. Meakin GH. Role of the Jackson Rees T-piece in pediatric anesthesia [editorial]. *Pediatric Anesthesia* 2007;**17**:613–15.
3. Hodges SC, Hodges AM. A protocol for safe anaesthesia for cleft lip and palate surgery in developing countries. *Anaesthesia* 2000;**55**: 436–41.
4. Malan CA, Wilkes AR, Hall JE, Gildersleve C. An evaluation of the filtration performance of paediatric breathing system filters at low flows. *Anaesthesia* 2007;**62**:504–8.
5. Cook TM. Novel airway devices: spoilt for choice? [editorial]. *Anaesthesia* 2003;**58**:107–10.
6. Bridgland IA. Monitoring medical devices: the need for new evaluation methodology [editorial]. *Br J Anaesth* 2001;**87**:678–80.
7. Grant LJ. Regulations and safety in medical equipment design [editorial]. *Anaesthesia* 1998;**53**: 1–3.
8. Brain AI. An evaluation of the laryngeal mask airway during routine paediatric anaesthesia [letter]. *Paediatr Anaesth* 1995;**5**:75.
9. Kelly F, Sale S, Bayley G, Cook T, Stoddart P, White M. A cohort evaluation of the pediatric ProSeal laryngeal mask airway in 100 children. *Pediatric Anesthesia* 2008;**18**:947–51.
10. Sanders JC, Olomu PN, Furman JR. Detection, frequency and prediction of problems in the use of the proseal laryngeal mask airway in children. *Pediatric Anesthesia* 2008;**18**:1183–9.
11. Rendell-Baker L, Soucek DH. New pediatric facemasks and anesthetic equipment. *Br Med J* 1962;**I**:1690.
12. Fisher DM. Anesthesia equipment for pediatrics. In: Gregory GA, editor. *Pediatric anesthesia*. New York: Churchill Livingstone; 1989. p. 437–75.
13. Malhotra N. Tracheal intubation in paediatrics: select tubes from the same manufacturer [letter]. *Anaesthesia* 2006;**61**:725–6.
14. James I. Cuffed tubes in children. *Paediatrics in Anaesthesia* 2001;**11**: 259–63.
15. Weiss M, Gerber AC. Cuffed tracheal tubes in children-things have changed [editorial]. *Pediatric Anesthesia* 2006;**16**:1005–7.
16. Khalil SN, Mankarious R, Campos C, Chuang AZ, Lemak NA. Absence or presence of a leak around tracheal tube may not affect postoperative croup in children. *Paediatr Anaesth* 1998;**8**:393–6.
17. Allt JE, Howell CJ. Down's syndrome. *British Journal of Anaesthesia. CEPD Reviews* 2003;**3**:83–6.
18. Gupta K, Harry R. Cutting paediatric tubes – a potential cause of morbidity [letter]. *Br J Anaesth* 1997;**78**:627.
19. Mackway-Jones K, Molyneux E, Phillips B, Wieteska S. *Advanced paediatric life support*. 3rd ed. London: BMJ Books; 2001.
20. Gursoy F, Algren JT, Skonsby BS. Positive pressure ventilation with the laryngeal mask airway in children. *Anesth Analg* 1996;**82**: 33–8.
21. Beveridge ME. Laryngeal mask anaesthesia for repair of cleft palate. *Anaesthesia* 1989;**44**:656–7.
22. Stocks RM, Egerman R, Thompson J, Peery M. Airway management of the severely retrognathic child: use of the laryngeal mask airway. *Ear Nose Throat J* 2002;**81**:223–6.
23. Auden SM, Lerner GM. Blind intubation via the laryngeal mask: a word of caution [letter]. *Paediatr Anaesth* 2000;**10**:452.
24. Sanders JC, Olomu PN, Furman JR. Detection, frequency and prediction of problems in the use of the proseal laryngeal mask airway in children. *Pediatric Anesthesia* 2008;**18**:1183–9.
25. Bagshaw O. The size 1.5 laryngeal mask airway (LMA) in paediatric anaesthetic practice. *Paediatr Anaesth* 2002;**12**:420–3.
26. Frohlich D, Schwall B, Funk W, Hobbhahn J. Laryngeal mask airway and uncuffed tracheal tubes are equally effective for low flow or closed system anaesthesia in children. *Br J Anaesth* 1997;**79**: 289–92.
27. Keidan I, Berkenstadt H, Segal E, Perel A. Pressure versus volume-controlled ventilation with the laryngeal mask airway in paediatric patients. *Paediatr Anaesth* 2001;**11**: 691–4.
28. Ozlu O, Turker AK, Ozgun G, Soykan I. Distal oesophageal pH measurement in children during general anaesthesia using the laryngeal mask airway, tracheal tube and face mask. *Paediatric Aneasthesia* 2001;**11**:425–30.
29. Baraka A, Choueiry P, Medawar A. The laryngeal mask airway for fibreoptic bronchoscopy in children. *Paediatr Anaesth* 2005;**5**:197–8.
30. Ong M, Chambers MA, Hullett B, Erb TO, von Ungern-Sternberg BS. Laryngeal mask airway and tracheal tube cuff pressures in children: are clinical endpoints valuable for guiding inflation? *Anaesthesia* 2008;**63**:738–44.
31. Wilkes AR, Hodzovic I, Latto IP. Introducing new anaesthetic equipment into clinical practice [editorial]. *Anaesthesia* 2008;**63**: 571–5.
32. Thomas PB, Parry MG. The difficult paediatric airway: a new method of intubation using the laryngeal mask airway, Cook airway exchange catheter and tracheal intubation fibrescope. *Paediatr Anaesth* 2001; **11**:618–21.
33. Hasan MA, Black AE. A new technique for fibreoptic intubation in children. *Anaesthesia* 1994;**49**: 1031–3.
34. Ellis DS, Potluri PK, O'Flaherty JE, Baum VC. Difficult airway management in the neonate: a simple method of intubating through a laryngeal mask airway. *Paediatr Anaesth* 1999;**9**:460–2.
35. Wrigley SR, Black AE, Sidhu VS. A fibreoptic laryngoscope for paediatric anaesthesia. *Anaesthesia* 1995;**50**:709–12.

36. Bein B, Wortmann F, Meybohm P, Steinfath M, Scholz J, Dörges V. Evaluation of the pediatric Bonfils fibrescope for elective endotracheal intubation. *Pediatric Anesthesia* 2008;**18**:1040–4.
37. Ravishankar M, Kundra P, Agrawal K, Kutralam NS, Arun N, Vijaykumar OP. Rigid nasendoscope with video camera system for intubation in infants with Pierre Robin sequence. *Br J Anaesth* 2001;**87**:728–31.
38. Depierraz B, Ravussin P, Brossard E, Monnier P. Percutaneous transtracheal jet ventilation for paediatric endoscopic laser treatment of laryngeal and subglottic lesions. *Can J Anaesth* 1994;**41**:1200–7.
39. Seefelder C, Kenna MA. Don't rely on the surgical airway: a case of impossible tracheostomy [letter]. *Paediatr Anaesth* 2000;**10**:224.
40. Hatch DJ. Paediatric anaesthetic equipment. *Br J Anaesth* 1985;**57**: 672–84.
41. Brown TCK, Fisk GC. *Anesthesia for children*. 2nd ed. Oxford: Blackwell Scientific Publications; 1992. p. 53–76.
42. Dhara SS, Pua HL. A non-occluding bag and closed scavenging system for the Jackson Rees modified T-piece breathing system. *Anaesthesia* 2000;**5**:450–4.
43. Orlikowski CE, Ewart MC, Bingham RM. The Humphrey ADE system: evaluation in paediatric use. *Br J Anaesth* 1991;**66**:253–7.
44. Rasch DK, Bunegin L, Ledbetter J, Kaminskas D. Comparison of circle absorber and Jackson Rees systems for paediatric anaesthesia. *Can J Anesth* 1988;**35**: 25–30.
45. Revell DG. A circulator to eliminate mechanical dead space in circle absorption systems. *Can J Anesth* 1959;**6**:98–103.
46. Rasch DK, Bunegin L, Ledbetter J, Kaminskas D. Comparison of circle absorber and Jackson Rees systems for paediatric anaesthesia. *Can J Anaesth* 1988;**35**:25–30.
47. Meakin GH. Low-flow anaesthesia in infants and children. *Br J Anaesth* 1999;**83**:50–7.
48. Brock-Utne JG. Humidification in paediatric anaesthesia [editorial]. *Paediatr Anaesth* 2000;**10**:117–19.
49. Heinsen A. A phylogenetic analysis elucidating a case of patient-to-patient transmission of hepatitis C virus during surgery. *J Hosp Infect* 2000;**46**:309–13.
50. Lawes EG. Hidden hazards and dangers associated with the use of HME/filters in breathing circuits. Their effect on toxic metabolite production, pulse oximetry and airway resistance. *Br J Anaesth* 2003;**91**:249–64.
51. Chau A, Kobe J, Kalyanaraman R, Reichert C, Ansermino M. Beware the airway filter: deadspace effect in children under 2 years. *Pediatric Anesthesia* 2006;**16**:932–8.
52. Bell GT, Martin KM, Beaton S. Work of breathing in anesthetized infants increases when a breathing system filter is used. *Pediatric Anesthesia* 2006;**16**:939–43.
53. The Medicine and Healthcare Products Regulatory Agency. Breathing system filters: an assessment of 104 breathing system filters. *www.mhra.gov*. March 2004.
54. Cann C, Hampson MA, Wilkes AR, Hall JE. The pressure required to force liquid through breathing system filters. *Anaesthesia* 2006;**61**:492–7.
55. Rees LM, Sheraton TE, Modestini C, Wilkes AR, Hall JE. Assessing the efficacy of HME filters at preventing contamination of breathing systems. *Anaesthesia* 2007;**62**:67–71.
56. Pirotte T, Veyckemans F. Ultrasound-guided subclavian vein cannulation in infants and children: a novel approach. *Br J Anaesth* 2007;**98**:509–14.
57. National Institute for Clinical Excellence. *Guidance on the use of ultrasound locating devices for placing central venous catheters*. London: NICE; 2002.
58. Kam PCA, Kavanaugh R, Yoong FFY. The arterial tourniquet: pathophysiological consequences and anaesthetic implications (review article). *Anaesthesia* 2002;**56**:534–45.
59. Crossley AWA, Holdcroft A. Physiology of heat balance. In: Jones RM, editor. *Royal College of Anaesthetists newsletter* 1993;**47**: 155–8.
60. Giesbrecht GG, Ducharme MB, McGuire JP. Comparison of forced-air patient warming systems for perioperative use. *Anesthesiology* 1994;**80**:671–8.
61. Cole R. When every second counts. *Child Health* 1995;**3**:63–7.
62. Rawlinson SC, Short JA. The representation of anaesthesia in children's literature. *Anaesthesia* 2007;**62**:1033–8.

FURTHER READING

Bingham R, Lloyd-Thomas A, Sury M, editors. *Hatch and Sumner's textbook of paediatric anaesthesia*. 3rd ed. London: Hodder Arnold; 2007.

Hatch D, Fletcher M. Anaesthesia and the ventilatory system in infants and young children. *Br J Anaesth* 1992;**68**: 398–410.

Steven JM, Cohen DE, Sclabassi RJ. Anesthesia equipment and monitoring. In: Motoyama EK, Davis PJ, editors. *Smith's Anesthesia for infants and children*. 6th ed. New York: Mosby; 1996. p. 229–79.

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Chapter 13

Equipment for regional anaesthesia

Susanne Krone

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The key to successful 'conduction anaesthesia' is deposition of the local anaesthetic drug accurately around the target nerve.

The introduction and widespread adoption of ultrasound imaging represents the largest change in regional anaesthesia in decades. Consequent to the use of ultrasound, for the first time the anaesthetist is able to view an image of the nerve directly, guide the needle under real-time observation, navigate away from sensitive anatomy, and observe the spread of the injected drug.

Previously, electrical stimulation, paraesthesia and the occasional unique anatomical feature such as the epidural loss of resistance to injection or detection of cerebrospinal fluid, were all that was available to confirm targeting. All such approaches to nerve targeting have relied on good anatomical knowledge and surface landmark identification. However, landmark techniques have limitations: variations in anatomy, the skill and experience of the anaesthetist, and equipment design all have an effect on success rates and complications. Many other regional anaesthetic (RA) blocks have relied on less precise information.

Equipment for RA is here broadly considered under two parts: devices concerned with target identification, and needle systems for drug delivery to the target area (including stimulating needles and their catheters).

NERVE LOCATION DEVICES

Ultrasound

In 1978 La Grange et al[1] used Doppler ultrasound to assist a series of supraclavicular brachial plexus blocks. Kapral et al[2] described ultrasound-guided supraclavicular blockade in 1994. Since then there has been a year-on-year increase in related publications and a similar growth in clinical practice.

Over the past decade a large number of portable machines have become available. The very first machines had limited functionality and produced what is by comparison now such a poor image that one is tempted to wonder why practitioners and developers persisted with the technology. However, there are now any number of very good-quality portable or 'laptop' style machines available (manufacturers include SonoSite, Esaote and GE Healthcare). The machines may have Doppler, *tissue harmonics* and *multibeam* technology as standard, making them true alternatives to the traditional cart-based machines typically seen in radiology departments (Fig. 13.1).

To optimize demonstration of nerves and surrounding structures, it is vitally important to understand the equipment and its limitations, and to have a good, sound anatomical knowledge of the structures being viewed.

Device specifics

Chapter 31 of this book is dedicated to the physics and technology of ultrasound imaging. The reader is encouraged to read the aforementioned chapter for

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Figure 13.1 M-Turbo portable ultrasound machine with linear broadband probe, SonoSite Inc. USA.

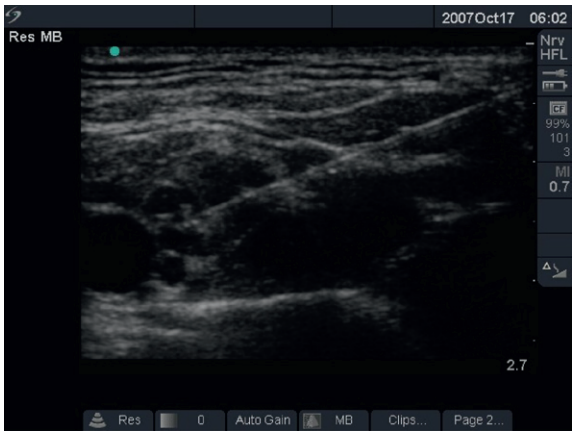


Figure 13.2 Screen grab, supraclavicular approach to brachial plexus. B-mode image. 0.5 cm markers and 2.7 cm total depth can be seen on the right of the screen.

further details of topics raised in this section on RA, which is concerned more with the application of ultrasound in RA and the small portable machines used for that purpose.

Most ultrasound machines have the following components in common:

- a signal generator producing an applied voltage (typically up to 100 volts) in brief bursts
- a piezoelectric transducer which converts electrical energy to acoustic pulses and vice versa
- a signal receiver which detects, amplifies and compresses signals returning to the transducer
- signal processing software for displaying the data in different modes
- a display together with the interface for controlling the machine
- a memory which stores still and video images and has connectivity ports for the output of images and other data.

Imaging modes used in ultrasound-guided regional anaesthesia

Modern scanners display ultrasound data in various forms. At present, *brightness* or B-*mode* is most commonly used for nerve imaging, producing a single two-dimensional image (hence also called *2D mode*) from a slice of approximately 2 mm thickness and adjustable depth (Fig. 13.2).

C-mode (colour) is useful to identify vessels within a region of interest (ROI) and flow is encoded as a colour image superimposed on the B-mode greyscale image. *Colour flow ultrasound* (CFU) measures directional flow and velocity whereas *colour power Doppler* (CPD) gives no directional information or estimation of velocity, but has greater sensitivity when the angle of the incident beam to the target surface is approaching the 90° angle – this being the angle that provides the best working image of target nerves.

These and other imaging modes, together with some further developments in image processing, are discussed more fully in Chapter 31. Software-driven post processing of the image is also now available to enhance needle visualization on some machines.

The recent advances in the development of three- and four-dimensional ultrasound imaging, although not yet routinely available, promise simultaneous multiple planes of view or a representation of the whole area of interest. This may give the operator an improved spatial awareness and understanding of anatomy and needle position.

Controls

On the ultrasound machine, the most important controls that are required to generate and optimize the image are those for frequency, depth adjustment and focal zone, time gain compensation (TGC) and imaging mode selection (2D, C-mode, etc.) (Fig. 13.3).

The selection of ultrasound frequency is a balance between obtaining the best resolution and being able to achieve adequate penetration of tissues. The higher the frequency of the ultrasound, the shorter the wavelength of the sound waves produced and the better the resulting image resolution. High frequency provides high-resolution images of superficial structures (brachial plexus), but poorer quality images of deeper structures. Lower frequencies permit better penetration of tissues and are necessary for imaging and needling deeper structures such as the lumbosacral plexus and neuraxial components.

Frequency adjustment facilities (presented sometimes as a choice of *penetration*, *general* and *resolution* modes) can be integrated into the system if using *broadband* transducers



Figure 13.3 Control panel of M-Turbo, showing rotary controls for gain settings and hard keys also for mode selection and depth settings as well as track pad and QWERTY keyboard for other functions. Soft keys (controlling varied functions) are visible as unlabelled buttons below the screen area.

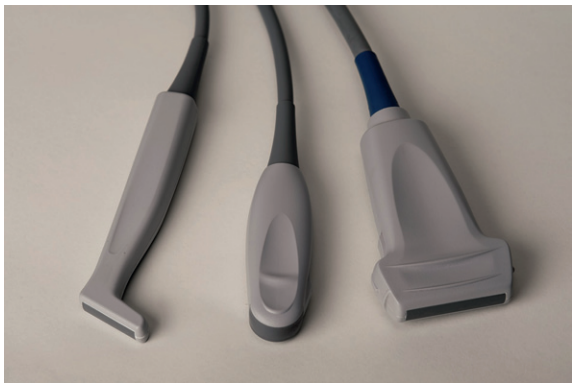


Figure 13.4 SonoSite transducers: from left to right: hockey stick (25 mm) high frequency linear array used in paediatric applications, small foot print sector probe (curvilinear array) as used for praecordial echocardiography, 38 mm high-frequency linear array used for peripheral nerve blocks.

Table 13.1 Different types of probe and their applications in anaesthesia

PROBE	TRANSDUCER ARRAY AND TYPICAL SIZE	FREQUENCY	NOMINAL FIELD DEPTH (MAX.)	RESOLUTION	BLOCKS
Linear	Linear 38 mm	6–13 MHz	6 cm	0.5 mm axial 1 mm lateral	Brachial plexus, abdominal wall, femoral and distal sciatic, peripheral nerves, vascular access
Hockey stick	Linear 25 mm	6–13 MHz	6 cm	As above but paediatric applications (small footprint)	
Sector	Curved 60 mm	2–5 MHz	30 cm	2 mm axial 3 mm lateral	Neuraxial, lumbar plexus, proximal sciatic

or it may require a change in transducer (probe) depending on the type of system in use.

The choice of correct depth setting places the target in the centre of the image, allowing the target and surrounding structures to be viewed optimally. Some ultrasound machines (SonoSite) are fitted with an auto-focus function in the middle of the image area, others allow adjustment of the focal zone.

Time gain compensation (TGC) adjusts the brightness of the image, excessive gain can obscure important structures and insufficient gain can result in missing structures of low reflectivity. TGC control can be adjusted via *sliders*, *near and far field* controls or *auto-gain* control, depending on the machine.

Transducers

Transducer characteristics, such as operating frequency and probe shape, determine the image generated. The transducer frequencies used for peripheral nerve blocks range from 3 to 15 MHz. Modern transducers are broad bandwidth (*broadband*) transducers that are designed to generate more than one frequency. For example, a SonoSite HFL38 6–13 MHz transducer can generate ultrasound ranging in frequency from 6–13 MHz and is a 38 mm sized linear probe. With broad bandwidth transducers, the operator selects the examination frequency to match the target requirement. Linear and *curvilinear* (curved) transducers are most useful for nerve imaging to provide high-resolution images. Linear arrays produce images with a finely sampled, rectangular field of view, whereas curved arrays produce a diverging sector-shaped field of view that expands beyond the lateral extent of the transducer (Fig. 13.4).

The probe used should match the procedure being performed (Table 13.1). Choosing the wrong probe can make identification of anatomy difficult. It is important to use the highest frequency probe available for the depth of the target being scanned.



Figure 13.7 A Braun Stimulplex peripheral nerve stimulator.

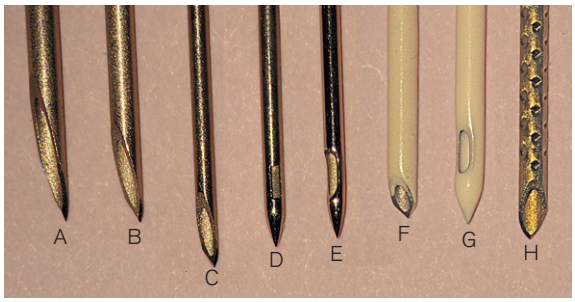


Figure 13.8 Needle tips for comparison: **A.** 21 G hypodermic needle. Spinal needles: **B.** 22 G Quincke, **C.** 25 G Quincke, **D.** 25 G Whitacre, **E.** 25 G Sprotte. Peripheral nerve block needles: **F.** 21 G Locoplex (*Vygon, France*), Teflon coated, **G.** 22 G pencil point Polymedic, (*te me na SAS, France*), Teflon coated, **H.** 22 G SonoPlex (*Pajunk, Germany*), proprietary NanoLine coating over embossed pattern.

Characteristics of a nerve stimulator for localization of peripheral nerves include:[7]

1. adjustable constant current output from 0 to 5 mA. It should be easy for the operator or an assistant to adjust the delivered current
2. a short stimulation pulse. The shorter the stimulation pulse, the greater the ratio of the current required to stimulate when the needle tip is 1 cm away from the nerve compared to when the needle is immediately adjacent to the nerve. Most commercially available nerve stimulators produce monophasic square wave impulses of between 0.1 and 1.0 ms
3. clearly marked polarity. The needle should be connected to the cathode (−) and the anode (+) to the surface of the patient's skin
4. display showing the current being delivered.

NEEDLES AND CATHETERS

General considerations

Needles used to administer local anaesthetic drugs may be selected on the basis of one or more of the following features:

- needle tip design
- facility for electrical stimulation
- suitability for catheter placement
- needle diameter
- needle length.

Some of these aspects are considered individually below.

Needle tip design

For the regional anaesthetist needle tip design is mainly concerned with reducing the risk of nerve injury. Although a variety of designs are available, none confer absolute safety: it is perfectly possible to place any needle tip inside a nerve and inject local anaesthetic (LA) into it causing either temporary or long-lasting damage, or no damage at all.

Traditional hypodermic needles have a *cutting tip* designed for easy penetration of skin and tissues and are best avoided for peripheral nerve blocks. In contrast a *short bevelled* needle is more 'blunt' in use and is believed to give improved 'feel' of anatomical layers aiding identification of tissue planes by eliciting a 'pop' as the layers are penetrated (compare Fig. 13.8 elements A, F, H). Theoretically, a short bevelled (or *facetted*) needle can touch a nerve eliciting paraesthesia with less risk of nerve injury,[1] but there is some controversy. In general, it appears that long, flat bevels are more likely to cause nerve trauma,[8] but that the trauma will be more serious if a short bevel needle does make vigorous contact with a nerve.[9]

Pencil point needle tips (Fig. 13.8 elements C, D and G) are believed to separate tissues through which they pass rather than cutting them. It is not clear whether a short bevel or a pencil point tip is safer to use.

The Huber tip provides a blunt relatively non-cutting and side-facing advancing edge, which facilitates catheter placement and is hence commonly used on epidural needles and other needles for catheter-based peripheral nerve block techniques (Fig. 13.9).

Stimulating needles

The shaft of a stimulating needle is coated with an insulator (e.g. Teflon) so that any electrical current passed through the needle spreads into the tissues from the tip only (Fig. 13.8 elements F–H). This coating also decreases friction allowing easier advancement of the needle. They now always have an electrical lead attached for connection



Figure 13.9 Tuohy type nerve block needle with Huber point. Note the white insulation material on the shaft.

to a nerve stimulator (in the past needles were used with a crocodile clip onto the non-insulated needle shaft).

Peripheral nerve block needles are usually supplied with a short length of clear flexible tubing between the needle hub and the Luer syringe connector in order to assist detection of aspirated blood and allow an '*immobile needle*' before and during drug injection.

Catheters

Insertion

Catheters are used to facilitate continuous nerve blockade. The different catheter sets available contain an introducer, a catheter and associated parts such as a filter and detachable connector. A popular introducer design is the 18 G insulated stimulating needle with a curved Huber tip (Fig. 13.10). It is believed to ease the placement of the associated 21 G catheter, particularly if the nerve is to be approached at right angles to its long axis. Another introducer design is the plastic over the needle cannula (usually 15–17 G) familiar as the traditional intravascular cannula, which after removal of the introducing needle is used to convey the catheter. This type of introducer for catheter placement tends to be more troublesome and less popular as once the needle has been removed, the cannula is prone to misplacement and kinking during insertion and passage of the final catheter.

Although a variety of different products are available and injection of LA or saline might open up the space to aid the insertion of the catheter, it can still be very difficult to position the catheter into the desired location.

Design

All catheters have length markings to allow calculation of the depth of insertion distal to the needle. Catheters may have a single end hole or multiple side holes. The latter are perhaps unsuitable for continuous peripheral nerve block, because one or more of the holes may lie outside the fascial plane in which the nerve lies. Infused local anaesthetic will take the path of least resistance, which may be through an orifice that is either some distance

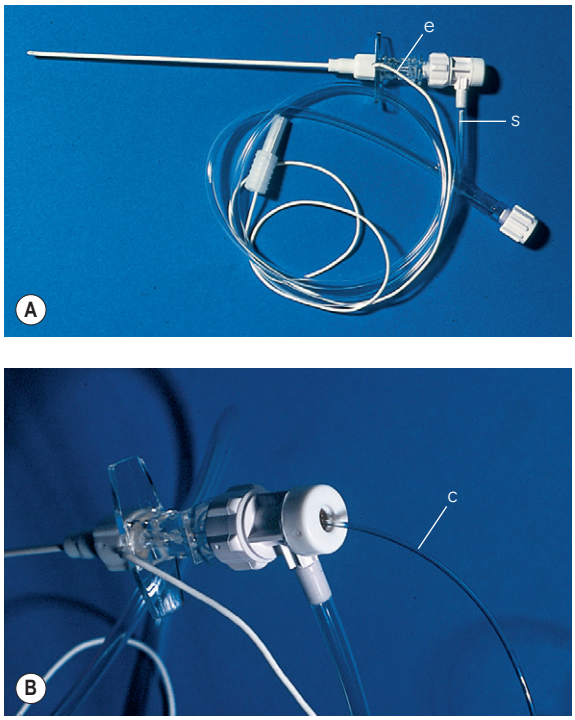


Figure 13.10 **A.** An insulated Tuohy type needle with an integrated electrical connection (E) and a side arm injection port (S). **B.** A catheter (C) can be seen entering the hub assembly ready to be passed through the needle.

from the nerve or outside its fascial surrounds. This leads to *secondary block failure* (the initial block which was administered through the needle works, but then wears off as the infusion fails to maintain anaesthesia).

Some very fine catheters have a removable wire stiffener inside which makes passage between the fascial planes easier. Unfortunately, this may also allow the catheter to pass *out* of the correct plane more readily, resulting in secondary failure of the block. Some catheters are radiopaque, but if not, contrast may be injected to confirm accurate placement.

A stimulating catheter is made from insulating plastic material and usually contains a metallic wire, inside which the current is conducted to its exposed tip electrode (Fig. 13.11). Stimulating catheters are placed through a nerve block needle, which itself may be placed using nerve stimulation. The catheter can then be stimulated to reconfirm the catheter tip position in close proximity to the target nerve.

Needle diameter and length

Needle diameters are usually quoted in terms of *standard wire gauge* – usually shortened to gauge and represented as G. Table 13.2 shows the metric equivalents of some

of the commonly used sizes. Standard notation is that the size is defined by the external diameter of the needle shaft, there being little or no standardization of internal diameters.

The diameter of needle required for a particular purpose will depend on two main factors:

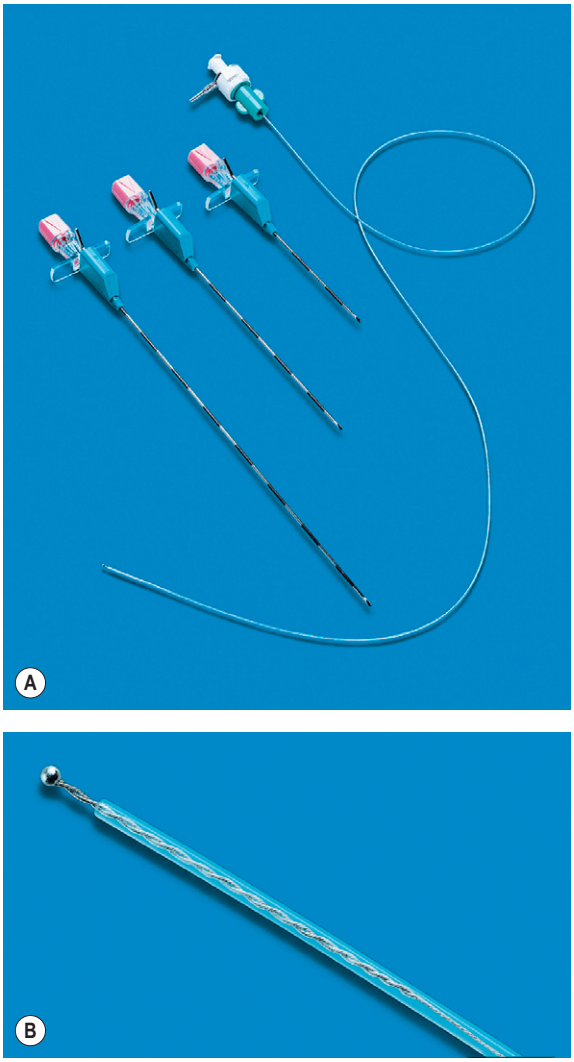


Figure 13.11 **A.** A Pandin stimulating catheter and selection of Huber tipped insertion needles. **B.** Magnified view of the catheter tip to show conducting parts. Photos courtesy of HDC Corporation, USA.

- *The viscosity of fluid that is to be injected*. Local anaesthetics are usually prepared in aqueous solutions, which will pass through small diameter needles, but needles which are very small will significantly limit the speed at which other fluids (e.g. CSF or blood) can be aspirated. The latter is required for confirmation of correct placement or for the early recognition of a potential complication. Oil-based agents (for neurolytic blocks) will only pass easily through the larger diameter needles (at least 18 G).
- *The rigidity required for its insertion*. The longer the needle the more flexible it becomes, making it more prone to deflection, bending, buckling or even breakage. Thus, epidural needles are usually of 16 G or 18 G size to minimize these risks. Spinal needles need to be even longer, but large diameter needles make post-dural puncture headache (PDPH) more likely and so finer needles are used and inserted through an introducer.

Given that shorter needles are easier to control; most manufacturers produce each needle design in a choice of lengths so that the shortest adequate length for the application can be selected.

Spinal anaesthesia

Spinal needles are used to administer intrathecal drugs. The main consideration in the design is the need to minimize the risk of PDPH. The needle tip should produce minimal trauma and make the smallest possible hole into the dura. The tip may be either a bevelled cutting design or a non-cutting rounded pencil point.

The Quincke spinal needle has a cutting tip (similar to a hypodermic needle: compare Fig. 13.8 elements A–C). 27 G Quincke needles have an incidence of 2.5–3.5% of PDPH. It is often opined that to insert these such that the bevel is parallel to the largely longitudinal dural fibres, produces a smaller defect in the dura as less fibres are transected by the cutting edge of the needle.

For any given needle type, the smaller the bore the less likely the patient is to develop PDPH. Extremely fine 29 G needles are available, but are prone to bending at insertion and identification of back flow of CSF or blood is very slow and can be difficult. Currently needles of 24–27 G are most commonly used.

'Atraumatic' pencil point spinal needles like the Whitacre needle have a completely rounded non-cutting bevel with a solid tip, the opening being 2 mm proximal to the

Table 13.2 Standard wire gauge sizes and the metric equivalent diameter

27 G	26 G	25 G	23 G	22 G	21 G	19 G	18 G	16 G
0.42 mm	0.46 mm	0.51 mm	0.61 mm	0.71 mm	0.81 mm	1.1 mm	1.2 mm	1.6 mm

tip on the side. An adaptation of the Whitacre needle design is the Sprotte needle with a blunt ogival (bullet shaped) tip with an elongated lateral needle opening and a wider internal diameter for CSF flow (Fig. 13.8, elements D, E). These needle designs aim to push or stretch the dural fibres aside rather than cutting them, resulting in better dural closure after removal of the needle. They are associated with a significantly lower incidence of PDPH (0.8–1%).

Potentially the lateral opening of the needle may straddle the dura such that CSF can be aspirated, but injected local anaesthetic may be deposited into the epidural space.

There are concerns that pencil point needles have to be inserted further into the intrathecal space to ensure that the lateral hole (which is not at the tip of the needle) is within the CSF. The issue being that, if the needle is inserted above the level of L3/4, this may lead to the tip of the needle penetrating the conus medullaris of the spinal cord.

All spinal needles have a removable stylet to stiffen the needle and prevent possible coring of the skin, with resultant obstruction of the needle or contamination of the spinal space with epidermal tissue and skin bacteria. Introducer needles (18–20 G) are available to facilitate the insertion of smaller (25–29 G) and blunt spinal needles, and to avoid contact of the spinal needle with skin and subcutaneous tissue.

Spinal needles are available at various lengths to suit different sized patients. Currently in use are needles from 50 mm in length for paediatric patients, to 152 mm long needles for very obese patients, with 90 mm lengths most commonly used for the majority.

The design of the hub should facilitate CSF visualization; some smaller gauge needle hubs are designed to have a magnifying effect to assist early recognition of CSF backflow.

Microspinal catheters

Although epidural catheter systems may be used for continuous spinal anaesthesia, concerns about PDPH have led to the development of microspinal catheters. Intrathecal catheters offer great flexibility in allowing both the extent and duration of a spinal anaesthetic to be titrated precisely.

Various designs of needle and catheter system are available ranging from 20 G catheter through 18 G Tuohy needle, through to a 24 G catheter placed over a 29 G Quincke spinal needle which is then withdrawn on the end of a flexible stylet. Fig. 13.12 shows a typical 25 G microcatheter through 21G Sprotte needle system. Catheters as fine as 28–32 G are available.

Dural puncture with an 18 G Tuohy needle makes this procedure most suitable for use in the elderly who are less prone to PDPH. A stiff 20 G catheter introduced in the subarachnoid space may induce transient radicular irritation.

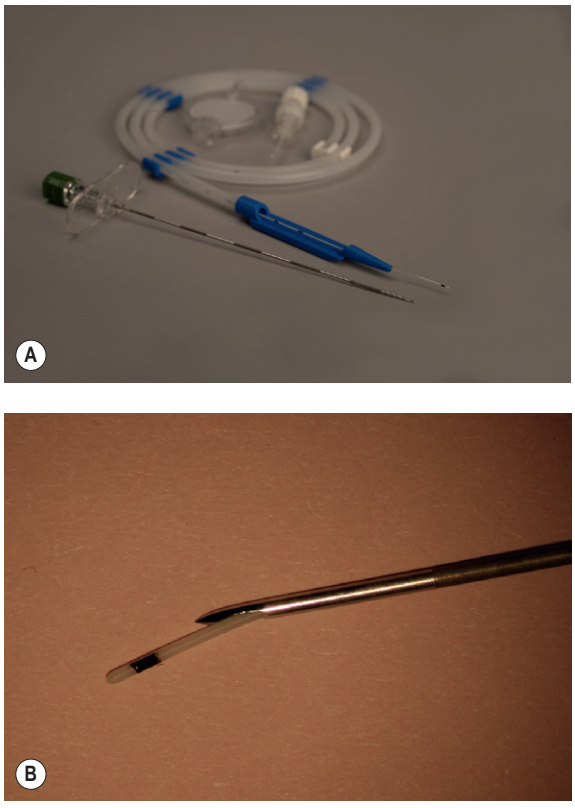


Figure 13.12 Intralong spinal catheter system, 21 G Sprotte needle with 25 G spinal catheter and internal Teflon coated stylet. Pajunk Medizintechnologie (Geisingen, Germany).

The utility of microcatheter systems is limited by the difficulties in ensuring that such a fine catheter is correctly placed. Combined spinal/epidural techniques provide many of the benefits without the same concerns regarding catheter position, leaving spinal catheter systems to a select, but loyal following. In the past, a small number of patients have developed the cauda equina syndrome following continuous spinal anaesthesia. Although initially ascribed to the use of the indwelling microspinal catheters themselves, it now seems likely that the true cause was the repeated localized exposure of nerve roots to very high concentrations of hyperbaric 5% lidocaine.[10]

Epidural anaesthesia

Epidural access is most commonly performed using the Tuohy needle with a Huber tip to facilitate insertion of the epidural catheter (Fig. 13.13). The Huber tip is relatively blunt and the shape ensures that the catheter emerges at an angle of 20° to the shaft. The disadvantage is that a catheter that has been passed beyond this needle tip cannot be withdrawn without the risk of being sheared off.

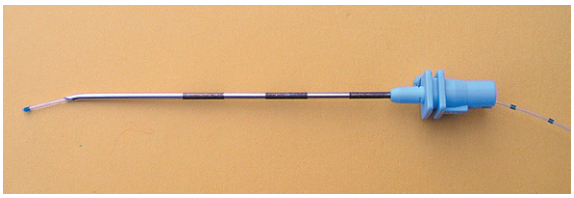


Figure 13.13 A Tuohy needle with a Huber tip and an epidural catheter emerging at an angle of 20° to the shaft.

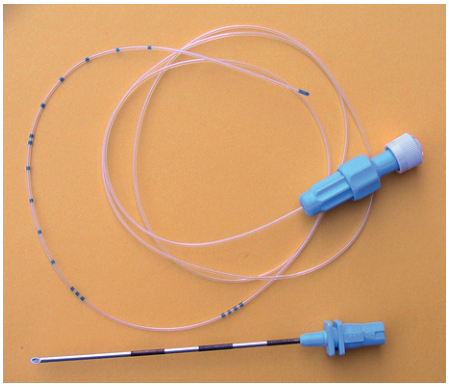


Figure 13.14 16 G Tuohy needle and associated catheter showing the centimetre markings.

The Crawford epidural needle, with a short conventional bevel, is sometimes used for a paramedian approach, as it slides off the lamina easily and a catheter can thread directly upwards when a 45–60° angle is used to enter the epidural space.

The standard Tuohy needle is approximately 11 cm in length with an 8 cm needle shaft and 3 cm hub assembly, although a longer version (15 cm) is available. The diameter is most commonly 16 or 18 G for adult patients.

Most needles have 1 cm markings on the shaft to show the depth of insertion from the skin surface (Fig. 13.14). A 19 G needle of 5 cm length for paediatric use has 0.5 cm surface markings. All epidural needles are supplied with close fitting stylets to prevent tissue entering the lumen on insertion. Catheters for epidural placement are offered with a choice of single end hole or multiple side holes, have appropriate centimetre markings and, once in place, are attached to a 0.2 µm mesh hydrophilic filter through which the anaesthetic agent is injected.

A low-friction syringe is most commonly used to aid identification of the epidural space using the technique of loss of resistance to injection. Such syringes can be made of plastic, glass or combined glass and metal. Spring-loaded and other novel loss-of-resistance devices are designed to automatically discharge when the epidural space is reached, but are not in common usage.

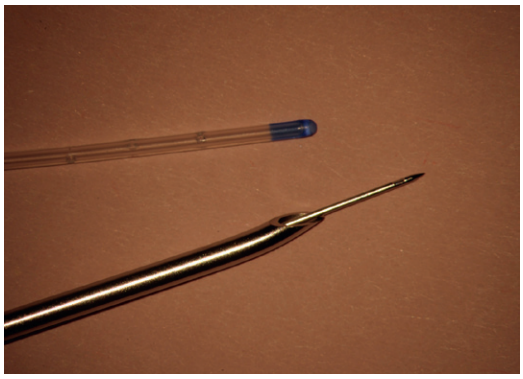


Figure 13.15 27 G pencil point needle exiting a 16 G Tuohy needle, and the associated epidural catheter for comparison.

Equipment for combined spinal/epidural (CSE) techniques

CSE combines the benefits of spinal anaesthesia – namely the certainty of a definitive endpoint (appearance of CSF) and rapid onset – with the flexibility of the catheter-based technique seen in continuous epidural anaesthesia.

The two approaches to CSE currently in practice are the needle-through-needle technique (NTN) and the double-space technique. When two separate spaces are used, the epidural component is completed before the intrathecal injection is attempted.

With the NTN, the intrathecal injection follows epidural insertion of the Tuohy needle, which then serves as an introducer for the very fine (25–27 G) intrathecal needle. After the spinal needle is removed the epidural catheter is inserted through the Tuohy needle. Usually, a pencil point type spinal needle of sufficient length (120 mm) to allow 13–15 mm projection beyond the end of a standard Tuohy needle is used (Fig. 13.15). Although sufficient projection is important longer spinal needles can puncture the anterior aspect of the dura and cause a greater CSF leak.

Commercially available kits have been produced for CSE to overcome various concerns (Fig. 13.16). Using matched needles to avoid damage to the spinal needle as it is passed through the Tuohy epidural needle prevents introduction of very fine metal particles into the CSF. Minimal drag between the two needles is essential to generate the dural 'click' when the spinal needle punctures the dura.

A conventional spinal needle, which does not lock within the much larger epidural needle, is minimally held by the dura alone and is difficult to handle and stabilize during injection of spinal medication. The displacement of the spinal needle during aspiration of the CSF and injection may result in failed anaesthesia or may push the spinal needle deeper, leading to nerve damage or anterior dural perforation. Several adjustable locking devices

recommendations for education and training in ultrasound-guided regional anesthesia. *Reg Anesth Pain Med* 2009;**34**:40–6.

7. Kaiser H, Niesel HC, Hans V. Fundamentals and requirements of peripheral electric nerve stimulation. A contribution to the improvement of safety standards in regional anesthesia. *Reg Anesth* 1990;**13**:143–7.
8. Selander D, Dhuner KG, Lundborg G. Peripheral nerve injury due to injection needles used for regional anesthesia. *Acta Anaesthsiol Scand* 1977;**21**:182.
9. Rice ASM, McMahon SB. Peripheral nerve injury caused by injection needles used in regional anaesthesia: influence of bevel configuration, studied in a rat model. *Br J Anaesth* 1992;**69**:433.
10. Faccenda KA, Finucane BT. Complications of regional anaesthesia Incidence and prevention. *Drug Safety: An International Journal of Medical Toxicology and Drug Experience* 2001;**24**(6):413–42.
11. National Patient Safety Agency. *Safer spinal (intrathecal) epidural and regional devices – Part A*. 24 November 2009. NPSA/2009/PSA004A (and updated 31 January 2011: available at http://www.nrls.npsa.nhs.uk/alerts/?entryid45=94529&q=0%C2%ACwrong+route%C2%AC accessed 28/7/11)

Chapter 14

Physiological monitoring: principles and non-invasive monitoring

Patrick T Magee

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INTRODUCTION

Monitoring the state of both the patient's physiology and the function of the anaesthesia delivery system is now an integral part of anaesthetic practice in the developed world. In some countries standards of monitoring are not enforced by law, yet it is difficult to prove that the absence of such monitoring is responsible for significant morbidity, since the trials required to prove this would be unethical. Anaesthesia has become safer in many countries over the last few decades, despite this fact it is difficult to demonstrate that the presence of multi-mode monitoring unequivocally reduces morbidity and mortality.[1] The cost of purchasing and maintaining such monitoring means it is unavailable to anaesthetists in many countries throughout the world and so the clinical monitoring skills of the anaesthetist remain crucial. Nonetheless, even in the developed world, all such standards emphasize the need for the continual presence of trained anaesthetic personnel.

The purpose monitoring is to advise the clinician of deviations from the normal and to warn of any unexpected, physiologically threatening events. In a study of closed claims in relation to deaths and cerebral damage under anaesthesia, two-thirds were found to be due to human error; of these, two-thirds were deemed to be due to problems with securing the airway, endotracheal intubation, ventilation and hypoxia.[2] A series of studies concluded that of the errors reported, 80% could be avoided by routine use of the pulse oximeter which, if used in conjunction with a capnometer, would lead to avoidance of 93% of such errors.[3] Fig. 14.1 shows a graphical representation of the development of a critical incident and avoidance of the harmful results which might follow.[4] Arguably, it is the function of monitors to help the anaesthetist detect a critical incident early and avoid development of injury to the patient.

There are a number of guidelines, published by various authorities such as the Association of Anaesthetists of Great Britain and Ireland (AAGBI), on minimum monitoring standards.[5] These help the clinical anaesthetist select monitoring appropriate to the circumstances. It is notable how much the AAGBI guidelines defer to the anaesthetists' own clinical judgement in this respect, since the ultimate 'monitors' are the anaesthetists themselves.[6]

CLASSIFICATION OF MONITORING EQUIPMENT

There are numerous ways of classifying monitoring equipment relevant to the anaesthetist. One way might specify the physiological system that it monitors: respiratory (including gas concentration, volume, flow rate and pressure); cardiovascular (e.g. ECG, arterial and venous pressures, cardiac output); neurological (e.g. EEG, neuromuscular junction); or metabolic (e.g. temperature or blood sugar). Another might include the degree of

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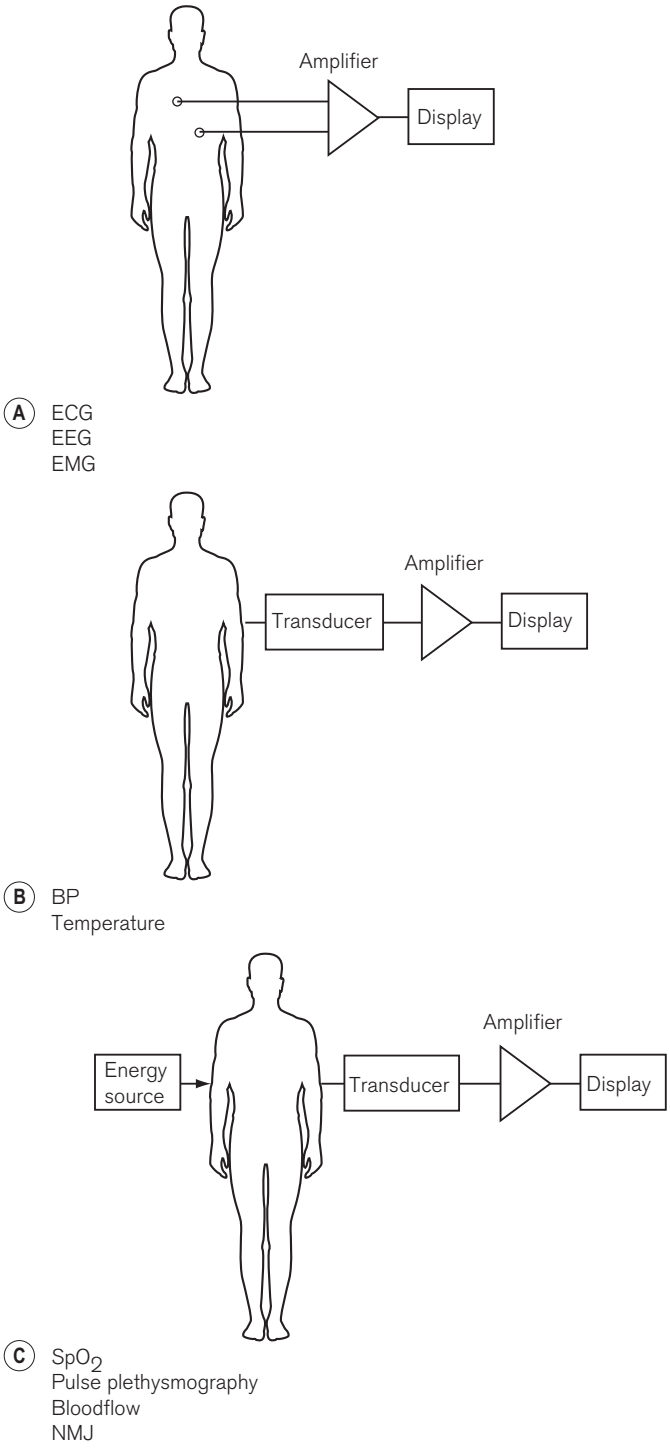


Figure 14.2 Basic classification of monitoring systems and some examples. **A.** Monitoring electrical signals generated by the patient. **B.** Conversion of measured variable to an electrical signal with a transducer. **C.** Passing energy through a patient and measuring the effect the patient has on it.

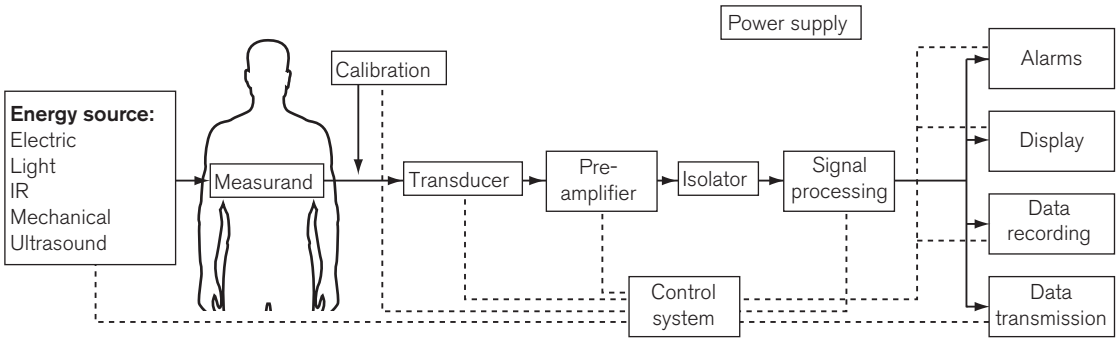


Figure 14.3 Generalized monitoring system showing the major components.

Table 14.1 The range of amplitudes and frequencies generated by biological potentials from different sources

SOURCE OF BIOLOGICAL ELECTRICAL POTENTIAL	AMPLITUDES	BANDWIDTH (Hz)
ECG	0.5–4.0 mV	0.01–250
EEG	5–300 μV	DC-150
EMG	0.1–5.0 mV	DC-10000

- The *signal to noise ratio* of the amplifiers should be high. This is the ability of an amplifier to ensure preferential amplification of the signal being measured in comparison to any electrical noise interfering with this process.
- The *common mode rejection ratio* should be high; this is achieved by having two input ports, one of which inverts the input signal, so that any random noise signals, such as inductively or capacitatively coupled signals, which are common to both ports, cancel each other out before entering the amplifier, while signal inputs, which are not common to both inputs to the same extent (such as the biological potential itself), are allowed to enter the amplifier. The difference between the two inputs is then subjected to electronic amplification. Fig. 14.4 shows single and double input amplifiers:
  - Where there are several amplifiers in a system, the *input impedance* of any single amplifier should be as high as possible so that the amplifier itself does not draw too much current from that being measured, thereby reducing its value.
  - The *output impedance* of an amplifier should be low so that the partially processed signal can be passed to the next stage with minimum attenuation.

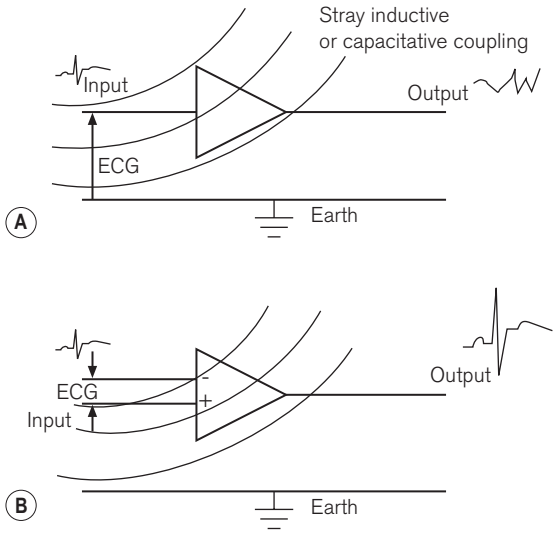


Figure 14.4 **A.** The output of a single input amplifier is adversely affected by stray signal coupling. **B.** The output of a differential input amplifier is relatively unaffected by stray signal coupling.

  - In order to process preferentially that component of the signal within the appropriate range of frequencies without attenuation or distortion and to eliminate inappropriate frequencies, the amplifier should have appropriate filtering in its circuitry to give adequate *bandwidth*.

Even surface electrodes must be carefully designed in order to minimize degradation of the small surface electrical potentials. A further requirement of the monitors is high-quality *electrical isolation* between any parts touching the patient, such as the electrodes, and any other electrical components within the device, which might lead to an electrical hazard to the patient. This is particularly important, since it is the only situation where there is a deliberate electrical connection made between the patient and a device which might be connected to a high-voltage mains

frequency source. If two pieces of electrical equipment are attached to a patient and one develops a fault allowing an unwanted earth pathway to occur, there is a risk of electrocution or diathermy burns. This is discussed in detail in Chapter 23.

The section below covers the electrocardiogram and monitoring of the neuromuscular junction, while Chapter 17 covers the electroencephalogram and its derivatives.

The electrocardiograph (ECG)

Despite the increasing intraoperative use of other monitors of cardiac function, such as ultrasound (see Chapter 16), the electrocardiogram remains extremely useful at detecting ischaemic events, providing close attention is paid to lead selection, amplification and filtering.[13] The electrical depolarization and repolarization of the myocardium is detectable on the surface of the skin, by the electrocardiogram in the form of the familiar PQRST complex. This can be recorded by the use of electrodes connected to limbs and chest, which look at the electrical vector of the ECG from slightly different points of view. The relationship between the electrical axis of the heart and its detection by different limb leads was described by Einthoven in 1901. The signal from the electrodes is then fed into an amplifier, which should meet the requirements discussed above.

As indicated in Table 14.1, the electrical potential from a surface ECG is in the range 0.5–4 mV, lying in the frequency range (bandwidth) of 0.01–250 Hz. The ECG is a complex waveform, which consists of a series of sinusoidal waves with different amplitudes, frequencies and phase relationships to each other. The ability of the monitor to process the ECG waveform depends on its ability to respond to the range of different frequencies of these sinusoidal components, with faithful, unattenuated and undistorted reproduction of the signal, and with the desired amplification or gain. This is a measure of the bandwidth of the monitor and the ECG monitor should ideally have the maximum required bandwidth indicated above. This is not practicable without also allowing amplification of noise within that bandwidth, thereby interfering with the ECG signal. Noise can originate from the amplifier circuit itself, from chest muscle electrical activity, from electromagnetic interference (inductive coupling), or from capacitative coupling to neighbouring electrical equipment such as diathermy apparatus (see Chapter 24). It is, therefore, more common to reduce the bandwidth to 0.05–100 Hz in a monitor with which the clinician wishes to make a range of diagnoses and to an even narrower bandwidth of 0.5–40 Hz in a monitor used in the operating room. The effect of this narrower bandwidth is to exclude high- and low-frequency components of electrical signal, which interfere with the ECG without adversely affecting its quality or the clinician's ability to diagnose ischaemia and arrhythmias.

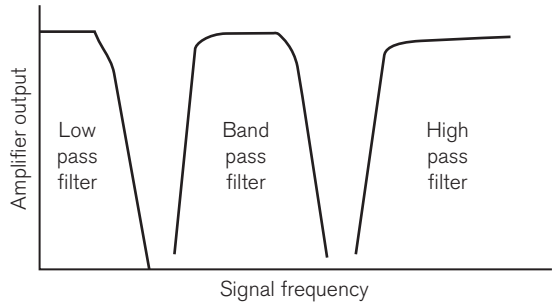


Figure 14.5 The effect of different electronic filters in allowing signals of different frequencies to pass through the amplifier.

Fig. 14.5 shows graphically the effect of electronic filtering, which is achieved by the addition of appropriate electronic components to an amplifier circuit. A low pass filter allows low-frequency components to pass through the amplifier, blocking high-frequency components. A high pass filter does the converse, allowing through high-frequency components. A band pass filter allows through a range of frequencies, blocking signals of frequencies above and below this bandwidth. There is a significant DC voltage of up to 25 mV at the skin–electrode interface, partly due to the resistance in the layers of the skin and partly due to the electrolytic reaction between the Ag/AgCl gel and metal component of the electrode assembly. An ECG amplifier has to eliminate this DC voltage as well as the high-frequency noise and, therefore, a band pass filter is appropriate.

Fig. 14.6 shows a block diagram of the components of a modern ECG monitor. As discussed above, the characteristic features include a differential input to ensure common mode rejection. Electrical isolation of the amplifier is also important for reasons discussed earlier, using either an isolation transformer with good insulation between primary and secondary windings, or an optical isolator, in which the output signal from the amplifier is converted by a light-emitting diode into a light signal and converted back into an electrical signal by a photo-detector, with good insulation between the two.

Even these design features may allow the ECG signal to be overwhelmed by high-power radio frequency diathermy signals. To deal with this, an additional technique is *adaptive noise filtering*. This is a digital electronic technique in which the noise signal is separately detected and digitally subtracted from the output signal, theoretically leaving a clean ECG signal. As Fig. 14.6 shows, analogue to digital conversion of the signal occurs to allow microprocessing of the signal. One of the many things a microprocessor provides is the storage, in digital form in the random access memory (RAM), of enough signal information to reproduce a screen width's worth of signal. This produces a more persistent and more easily readable screen than the previous phosphor dot screen. The RAM allows continual

BLOOD PRESSURE MONITORING

Monitoring blood pressure (BP) in the perioperative period has been routine since 1903 and is part of contemporary minimum monitoring standards. It is usually done non-invasively, except where there is a clinical need dictating 'beat to beat' invasive monitoring. Arguably, cardiac output would be a more useful measure of cardiovascular wellbeing, but has always been more complex to carry out than blood pressure measurement. Non-invasive blood pressure measurement became relatively easy to do early on in the history of clinical monitoring, but the earliest blood pressure measurements were invasive, using a manometer (see below). The manometer is still used in many settings, as a mercury manometer attached to a non-invasive sphygmomanometer cuff (see below) or as a water manometer attached to a central venous catheter to measure pressure directly. The manometer is a column of fluid, at the bottom of which the pressure is equal to the weight of fluid above it, divided by the cross-sectional area of the tube. Hence the pressure P is given by:

$$P = \frac{Weight\ of\ fluid}{Cross\text{-}sectional\ area\ A\ of\ tube} = \frac{\rho ghA}{A} = \rho gh$$

where ρ is fluid density, g is acceleration due to gravity, h is the height of fluid in the manometer column and A is the cross-sectional area of the tube. Hence the colloquial use of a column height (e.g. mmHg, cm H_2O) to describe pressure has evolved. The hydraulic pressure being measured is applied to the bottom of the manometer column or to one side of the 'U-tube' form of the manometer, shown in Fig. 14.7.

The venous circulation is a low-pressure system, and in the upright or partially upright patient, the venous pressure at the head will be significantly less than at the feet; this should be remembered in deciding where to site the pressure transducer, when measuring central venous pressure. This difference in arterial pressure from top to bottom may not be so apparent, because the arterial circulation is a high-pressure system with regionally adjustable resistance to flow.

Non-invasive arterial blood pressure measurement techniques include auscultation using a sphygmomanometer, the oscillotonometer, the oscillometer and the volume clamp method (Peñaz technique, see below).

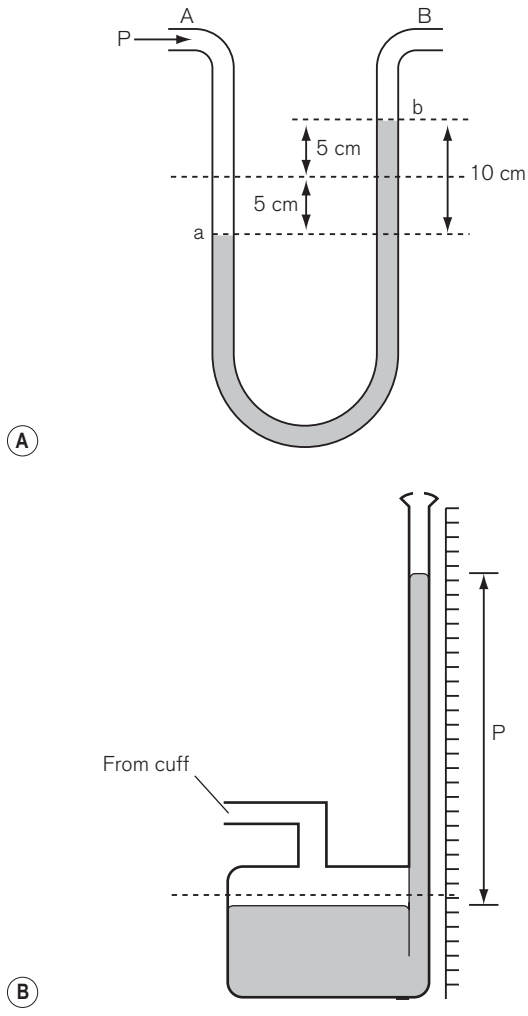


Figure 14.7 Two sorts of manometers to measure pressure: **A.** a U-tube differential manometer; **B.** a single tube manometer, of the type used in a sphygmomanometer.

Non-invasive arterial blood pressure (NIBP) measurement

NIBP methods all function by inflating an occlusive cuff around a limb to a pressure above the expected arterial pressure and, on subsequent gradual, stepwise or continuous deflation,[17] detecting the return of the pulse downstream of the cuff. Detection methods are described below in more detail. Most non-invasive methods are necessarily intermittent, although the volume clamp method of Peñaz (Finapres; see later) is continuous. In all non-invasive methods that use a cuff, the cuff size is important for accuracy; the width of the inflatable bladder of the cuff should be 40% of the mid-circumference of the limb concerned and its length should be twice this width. A cuff which is too narrow overestimates the blood pressure, while a cuff which is too wide or too loosely wrapped around the limb, underestimates it. Normally the cuff is wrapped round the upper arm to target the brachial artery, but if this is unsuitable (e.g. surgery on the arm or lymphoedema due to breast surgery), then the popliteal artery is targeted by wrapping the cuff around the calf, or, indeed, the posterior tibial artery if the cuff is wrapped around the ankle.[18]

The sphygmomanometer

The sphygmomanometer is a device that uses a combination of a pneumatic cuff (to wrap around the limb in which the arterial pressure is to be measured), an inflating bulb, a release valve, and a mercury manometer of the type shown in Fig. 14.7B. The classical method of pulse detection is *auscultation*. If the cuff is used on the upper arm, the sounds of the pulse returning are heard over the brachial artery. These sounds are known as *Korotkoff sounds* and their onset corresponds to the occurrence of turbulent flow in the artery as the cuff pressure falls below systolic blood pressure; the muffling or disappearance of these sounds corresponds to diastolic pressure.

The reason that Korotkoff sounds correspond to these pressures is unclear, but it is, therefore, not altogether surprising that it does not correlate well with invasive BP measurement under all circumstances. It has been shown that this method overestimates at low pressure and underestimates at high pressure, and has large inter-observer variation.[19] On the other hand, the method is simple, uses low-level technology, and has a long history of use. Other methods of pulse detection with the sphygmomanometer include palpation and ultrasound. Palpation is more prone to error in the presence of bradycardia or too rapid cuff deflation and it has been shown also that this method underestimates systolic pressure by 25%.[20]

The oscillotonometer

Fig. 14.8 shows the double cuff system used in oscillotonometry. The upper cuff performs the same function as that in the sphygmomanometer, namely limb occlusion. The lower cuff is wider and acts as the pulse detection system. Both cuffs are connected via an inflating bulb and air release valve to an airtight box containing two aneroid barometers. One aneroid (B_1) is relatively rigid and its inside is connected to atmosphere to measure the absolute pressure. The other aneroid (B_2) is relatively sensitive and its inside is connected to the distal cuff. Both aneroids are connected together by a lever, which is also toggled to the dial pointer. When the system is pressurized by manual inflation (using the bulb), both cuffs, the airtight box and the aneroid B_2 are filled to the same pressure. This causes the aneroid B_1 to be compressed and via its mechanical linkage moves a pointer to display the pressure in the system.

By switching on the release valve, a narrow diameter tube connecting the inside of aneroid B_2 to the atmosphere is opened and air is allowed to leak gradually out of the system. This allows deflation of the lower cuff to be marginally delayed behind the upper cuff. As the upper cuff pressure falls below systolic pressure, arterial pulsations impinge on the lower cuff and these are transmitted to aneroid B_2; with the release valve switch in this position, aneroid B_2 is also connected to the pointer and the pulsations, which are shown on the dial, increase significantly as systolic BP is reached. On releasing the valve switch, the pointer is reconnected to aneroid B_1 and the actual BP is indicated at this point. With the valve reswitched to air leak mode, the pulsation fluctuations gradually disappear as diastolic pressure is approached, and the actual pressure at which this occurs is once again read by reconnecting the pointer to aneroid B_1. Its accuracy is better at systolic pressure than diastolic.

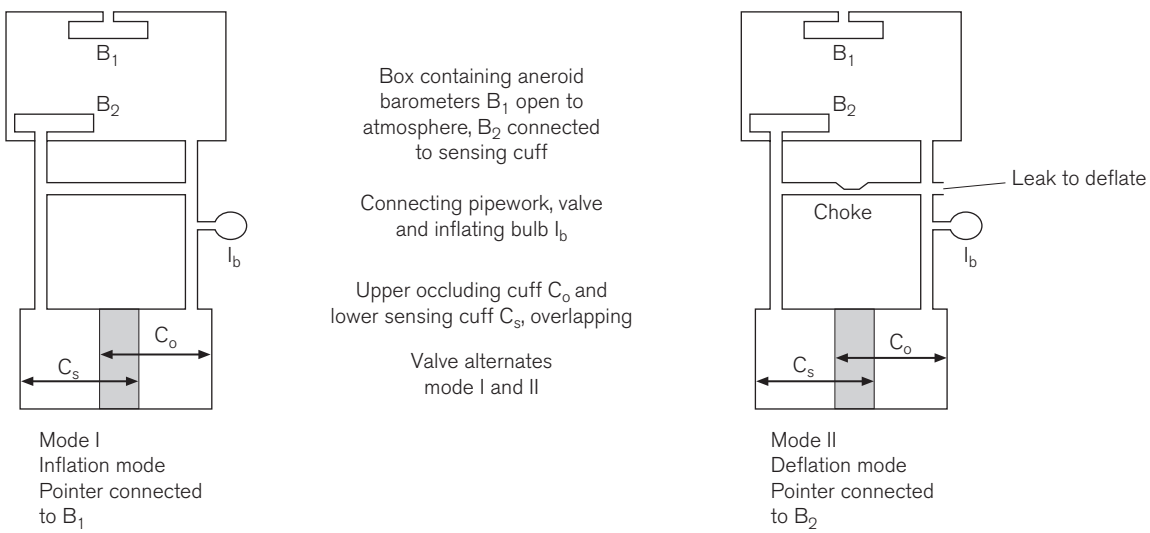


Figure 14.8 The working principle of an oscillotonometer.
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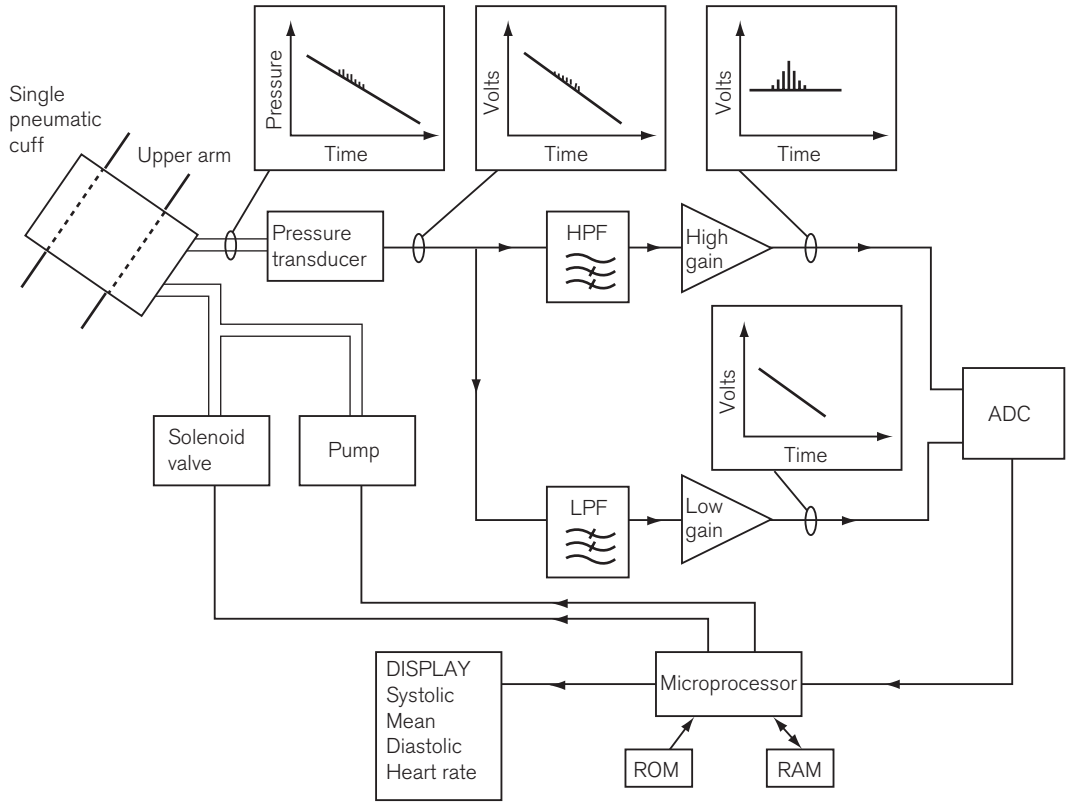


Figure 14.9 The components and working principles of an oscillometer.

Oscillometry

This method uses a single cuff that both compresses the limb and detects pulsations and uses hydraulics linked to modern electronics to obtain BP measurements. Some systems use a single tube for both inflation and detection, and others use two. Fig. 14.9 shows the components of the system. The hydraulic system consists of a pump to pressurize the system and a solenoid valve through which to depressurize the system, either in discrete steps or continuously, connected to a microprocessor. The detection system consists of a pressure transducer, whose output signal is fed through filters to amplifiers. On deflation, one part of the transducer signal goes through a high pass filter to pass only the high-frequency pulsation components to a high-gain amplifier that amplifies the pulsations due to the oscillations of the arterial wall.

The onset of the rapid increase of pulsations corresponds to systolic pressure, maximum pulsations to mean pressure, and the rapid offset of pulsations to diastolic pressure. The other part of the transducer signal goes through a low-pass filter and a low-gain amplifier to produce a signal proportional to the cuff pressure. The outputs from both amplifiers are passed through an analogue to digital converter to the microprocessor, which calculates systolic, mean and diastolic pressures. Depending on the exact algorithm being used, the microprocessor may also compare the measured systolic and diastolic values to the values calculated from the mean value, at which the device has the greatest accuracy, as the oscillations are maximal at mean pressure. The microprocessor also controls the pump and solenoid valve and connects to a display.

Peñaz volume clamp technique

This method depends on the hypothesis that if the transmural pressure of an artery (the difference between the externally and internally applied pressure across the arterial wall) is kept constant, then the diameter of the artery also remains constant, as will the volume of blood within it and the absorption of infrared light across its lumen. The device consists of a low compliance finger cuff and tubing connected to a rapidly responding solenoid valve and air pump. It also has a light emitting diode as an infrared light source, which transmits light through the finger; the transmitted light is detected on the opposite side by a photodetector. The pump, the solenoid valve and the microprocessor function together to keep the light transmission through the finger constant by maintaining

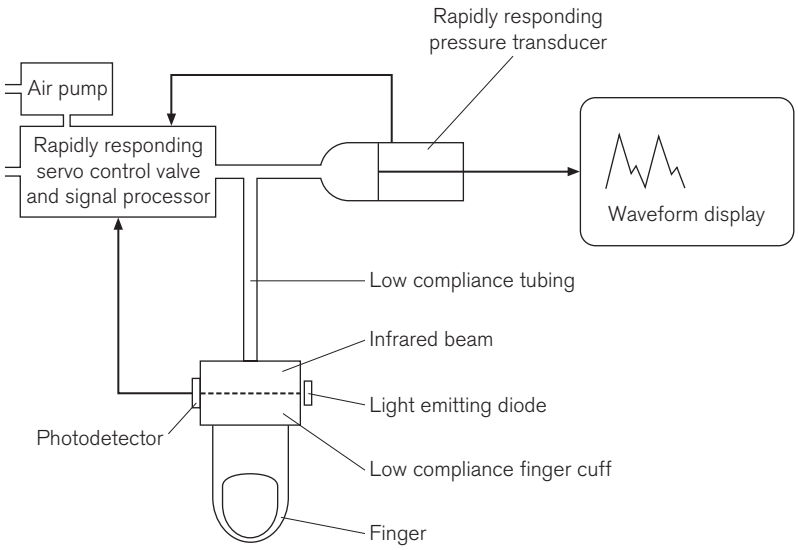


Figure 14.10 The components of the 'Finapres' volumetric clamp method for measuring blood pressure.

the transmural pressure. The device does this by altering the compression in the finger cuff throughout the cardiac cycle. At any instant the cuff pressure is the same as arterial pressure, and the low compliance of the system ensures rapid response. The output on the display therefore resembles an arterial trace. The *Finapres*, which is the manufacturer's name given to the device, is shown schematically in Fig. 14.10. There have been reports of varying accuracy[21,22] and it is no longer in extensive clinical use.

PULSE OXIMETRY

Principles

Table 14.2 shows that it is surprisingly difficult even for trained anaesthetists to ascertain accurately the patient's state of oxygenation, particularly where the oxygen saturations have fallen below optimal levels. Arguably pulse oximetry, first introduced in the early 1980s, has revolutionized clinical monitoring in this respect. It should not be thought of as a replacement for other oxygen monitors, such as those on anaesthetic workstations, but it does provide the best non-invasive monitor of patient oxygenation. The pulse oximeter uses two technologies: one is pulse plethysmography to detect a pulse waveform; the other is infrared spectroscopy to detect the absorption, by the tissue under the probe, of light at two wavelengths in the red and infrared wavebands.

Pulse plethysmography detects the cyclical change in volume of the artery as change in light absorption across it, an increase in one being an increase in the other. Red or infrared light is used for the same reasons as in clinical spectroscopy, namely that there are absorption spectra in this waveband across an artery. The absorption signal is then electronically processed, including amplification and display. In a modern pulse oximeter, this signal processing and display scaling occurs automatically, so the size of the plethysmographic signal cannot be taken as a quantitative indicator of pulse volume; a pulse volume below a certain threshold may not be detected and amplified at all.

Apart from the objective indication of SpO_2, pulse oximetry may be described as tending to 'fail-safe', in that the

Table 14.2 The ability of experienced anaesthetists to detect hypoxaemia

SPO_2 READING	PERCENTAGE OF ANAESTHETISTS DETECTING CYANOSIS
96–100	1.1
91–95	12.3
86–90	22.5
81–85	29.2
76–80	12.3
<75	22.5

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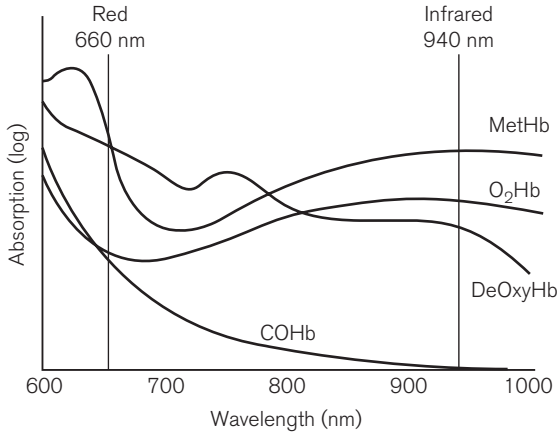


Figure 14.11 The absorption spectra in the red and infrared region for different species of haemoglobin.

technique fails and therefore alarms if the pulsatility of the pulse waveform decreases to below a critical level. Further information may be gained by the display of a plethysmographic trace, which enables the differentiation between the pulse waveform and artefacts. No pulse oximeter reading should be accepted unless the plethysmographic trace can confirm lack of artefact (see below, under Limitations).

The spectroscopic aspect of the technology is based on the absorption of light by a 'dye' in tissues, in which absorption is proportional to concentration of the dye and the path length through the tissue, and which also depends on the wavelength of light used; these factors are embedded in the *Beer-Lambert laws* (see Chapter 15). Fig. 14.11 shows the absorption spectra in the red and infrared wavebands of a number of haemoglobin species, including oxygenated and reduced haemoglobin, as well as the less common species, which can nevertheless unhelpfully contribute to a pulse oximetric signal.

The pulse oximeter has light-emitting diodes (LEDs), which transmit light alternately at two wavelengths: 660 nm (red) and 940 nm (infrared). It can be seen that at 660 nm, the absorption of reduced haemoglobin exceeds that of oxygenated haemoglobin and at 940 nm, the converse is true. The absorption scale in Fig. 14.11 is logarithmic, so the differences in absorption are large. The device uses two wavelengths, because it needs to distinguish between absorption of light in pulsatile and non-pulsatile tissues under the probe. Ideally the two wavelengths used should correspond to those at which absorptions of both haemoglobin species (reduced and oxygenated) are equal (the *isobestic point*) and at which they are furthest apart. LEDs of 660 and 940 nm are used because of their ease of production, constant outputs and narrow bandwidths.

The probe is put on a digit, an earlobe,[23] the bridge of the nose or the forehead[24] and the more peripherally placed it is, the longer the delay in the device detecting an event causing oxygenation change.[25] Probes should be appropriately designed for paediatric use[26] because of the *penumbra effect* where, because of the small size of the digit to which the probe is attached, the path length of the light at the two wavelengths differs significantly. The LED's rate of alternating on and off is 400 Hz, with a measurable 'off' phase, so the photodetector allows for ambient lighting. The photodetector measures the light transmission (transmission fraction = 1 – absorption fraction) of both pulsatile (AC) and non-pulsatile (DC) components at both wavelengths. The pulsatile components are assumed to come mainly from the arterial blood and non-pulsatile components from all other tissues including venous blood. The AC component of absorption represents 1–2% of the total, which puts significant demand on the accuracy of the device. Fig. 14.12A shows the raw signal, with different transmissions of each component at each wavelength. The first stage of signal processing is to change the amplitude of the signal at one of the wavelengths in order to equalize the DC components. The AC components, which are the signals of interest in this context, are now measureable against comparable denominators. The microprocessor in the device calculates the ratio of absorption of the AC component at 660 nm to the absorption of the AC component at 940 nm.

The microprocessor has stored in its memory a number of discrete values of oxygen saturation of arterial blood samples from healthy subjects, measured by a multi-wavelength co-oximeter. Fig. 14.12B shows this ratio plotted against the oxygen saturation and represents the calibration curve for the pulse oximeter. The term SpO_2 is used to describe the oxygen saturation derived from this curve. Note that an absorption ratio of 1.0 corresponds to a SpO_2 of 85%. The SpO_2 displayed is a value averaged over a number of beats, so a change in saturation may not immediately be displayed.

Limitations

Clearly, most values on the calibration curve shown in Fig. 14.12B are either interpolated or extrapolated. Since it would be unethical to collect blood from subjects who had been exposed to life-threatening hypoxia, it is to be expected that SpO_2 values below about 80% will be less accurate. In normal clinical circumstances, this is tolerable because values below 90% would be acted upon.

Pulse oximeters are designed to take account of two haemoglobin species, oxygenated and deoxygenated haemoglobin, to calculate *functional* saturation. As Fig. 14.11 shows, the presence of other haemoglobin species, such as methaemoglobin or carboxyhaemoglobin will affect the measured absorbances at the two wavelengths used. In particular, it can be seen that carboxyhaemoglobin resembles oxyhaemoglobin at 660 nm. This means that the pulse oximeter may give a falsely high SpO_2 in smokers,

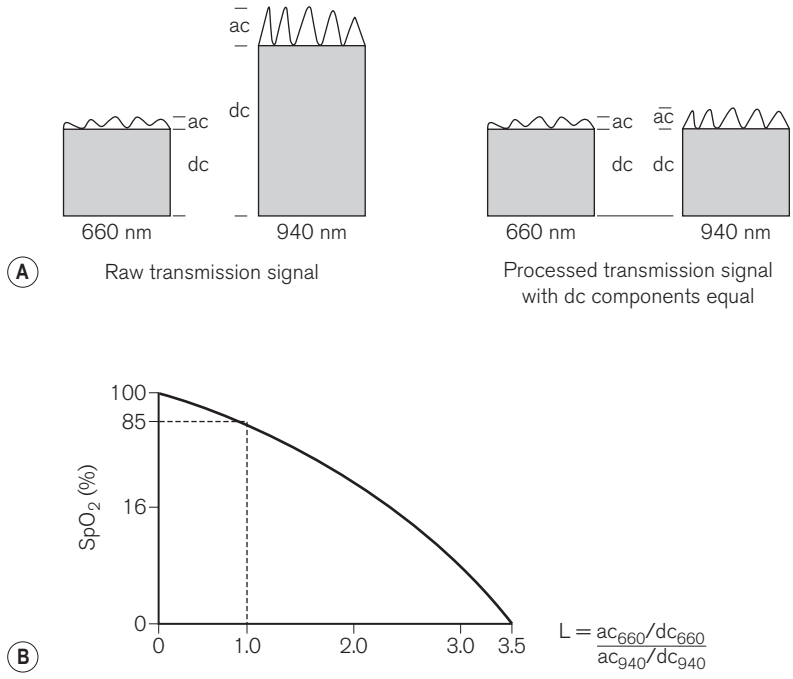


Figure 14.12 **A.** The crude AC and DC signals from a pulse oximeter probe are 'normalized' to make the DC components equal. **B.** The curve of absorption ratio to SpO_2.

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in whom COHb levels can reach 20%. Clearly a pulse oximeter should not be used to assess the oxygenation of a patient who has suffered from carbon monoxide poisoning. Similarly, methaemoglobin, caused by a number of drugs including local anaesthetics and nitrates, resembles deoxygenated haemoglobin at 660 nm. Only a co-oximeter with a minimum of four wavelengths can distinguish these four species, to calculate *fractional* saturation.

If vascular tone is markedly altered, then there is some limitation to the accuracy of pulse oximetery. This applies to hypertension or vasoconstriction induced by cold[27] or otherwise;[28] however, the plethysmographic pulse width has been shown to reflect changes in vascular resistance,[29] and ear pulse oximetry may be more useful in peripheral vasoconstriction.[23] Where vascular tone is reduced, such as in hypovolaemia or sepsis, a peripheral plethysmographic signal may not be detectable.[30] However, the quality of the waveform[31] may be useful in determining the clinical response to fluid resuscitation. If the venous circulation is hyperdynamic, as in tricuspid regurgitation,[32] or if the perivascular tissue is pulsatile,[33] then the oximeter will tend to underestimate arterial oxygenation values.

Foetal haemoglobin has the same properties of light absorption as adult haemoglobin within the wavebands being discussed, so the pulse oximeter should be as accurate in neonates as adults. Bilirubin does not absorb light significantly in the waveband of interest; therefore, jaundice does not affect the accuracy of the pulse oximeter. Both foetal haemoglobin and bilirubin, however, affect the accuracy of a multi-wavelength co-oximeter. Skin pigmentation does not usually affect accuracy, but some dark nail polish does. Intravenous dyes, such as methylene blue and indocyanine green, alter the absorption spectrum of haemoglobin in the range of the pulse oximeter, thus attenuating its accuracy.

Pulse oximeters are also prone to error in the presence of movement and vibration,[34] or electromagnetic interference from ambient light, diathermy or mobile telephones.[3]

It is inappropriate to compare SpO_2 measured by a pulse oximeter with the value of oxygen saturation derived from the oxygen dissociation curve following measurement of pO_2, pCO_2 and pH.

BODY TEMPERATURE MONITORING

At both physiological and biochemical levels, the human body functions optimally at 37 °C. Anaesthesia and surgery both militate against this by tending to allow body temperature to fall and recovery to be delayed after prolonged surgery (see also Chapter 30). On the other hand, malignant hyperthermia is a potentially fatal condition caused

by some anaesthetic drugs in patients pharmacogenetically predisposed to it. It is, therefore, essential for the anaesthetist to monitor body temperature.

A traditional way of measuring the temperature of a patient is to use a glass thermometer. This consists of a glass fluid-filled bulb attached to a calibrated glass tube. The glass bulb is placed against the tissue where temperature needs be measured, causing the fluid contained therein to heat up to the same temperature as the tissue. The resultant expansion of the fluid causes it to move into the calibrated glass tube as a column. The temperature can be read off the tube at the point where the head of the fluid column stops. A constriction is placed at the base of the tube so that when the bulb temperature drops and the bulb fluid contracts, the fluid column breaks allowing the final reading of the thermometer to be maintained.

Mercury is frequently the fluid used as its expansion characteristics allow it to cover a wide range of temperatures. Alcohol is also used, covering a narrower measurement range.

A *thermistor* is a semiconductor device whose electrical resistance changes with temperature in an almost linear fashion. It is the basis of both the nasopharyngeal temperature probe[35] and some tympanic membrane thermometers.

A *thermocouple* consists of two dissimilar metals, connected together at the *junction* or *measuring end* and completing an electrical circuit (the opposite ends form the *tail* or *reference end*). A potential difference is generated between the two ends proportional to the difference in temperature between them, although the relationship is a non-linear one. This is the *Seebeck effect*, and is the basis of thermocouple function.



Figure 14.13 A tympanic thermometer.

In the infrared tympanic thermometer (Fig. 14.13) a series of thermocouples (thermopile), detect the infrared radiation from the tympanic membrane.[36] The thermopile generates a potential difference proportional to tympanic membrane temperature.

Liquid crystal displays are made of materials that change colour with an alteration in temperature. They have been shown to demonstrate hysteresis and are sensitive to draughts.[37]

REFERENCES

1. Watkinson PJ, Barber VS, Price JD, Hann A, Tarassenko L, Young JD. A randomized controlled trial of the effect of continuous electronic physiological monitoring on the adverse event rate in high risk medical and surgical patients. *Anaesthesia* 2006;**61**:1031–9.
2. Utting JE. Pitfalls in anaesthetic practice. *BJA* 1987;**59**:877–90.
3. Runciman WB, Webb RK, Barker L, Currie M. The pulse oximeter: applications and limitations. The first 2000 AIMS reports. *Anaesth Intensive Care* 1993;**21**:543–50.
4. Schreiber P, Schreiber J. *Anesthesia systems risk analysis and risk reduction*. Telford PA: Dräger; 1987.
5. AAGBI. *Recommendations for standards of monitoring during anaesthesia and recovery*. 4th ed. March 2007.
6. Thompson JP, Mahajan RP. Monitoring the monitors. *Editorial. BJA* 2006;**97**:1–3.
7. Görges M, Staggers N. Evaluations of physiological monitoring displays; a systematic review. *J Clin Monit Comput* 2008;**22**:45–66.
8. *IEC 60601-1*, Medical electrical equipment—Parts 1-8: General requirements, tests and guidance for alarm systems in medical electrical equipment and medical electrical systems. Geneva: International Electrotechnical Commission; 2005.
9. Edworthy J, Hellier E. Alarms and human behaviour: implications for medical alarms. *BJA* 2006;**97**: 12–17.
10. Reich DL. Controlling data flow enhances anesthesiology's role in perioperative care. *J Clin Monit Comput* 2008;**22**:221–4.
11. Bitterman N. Technologies and solutions for data display in the operating room. *J Clin Monit Comput* 2006;**20**:165–73.
12. Smith RJ. *Circuits, devices and systems: a first course in electrical engineering*. 2nd ed. *Electronic amplifiers*. New York: Wiley; 1971. p. 343–90.
13. Mark JB. Cardiovascular anesthesiology: multimodal detection of perioperative myocardial ischemia. *Tex Heart Inst J* 2005;**32**:461–6.
14. McGrath CD, Hunter JM. Monitoring of neuromuscular block. *BJA Continuing Education* 2006;**6**:7–12.
15. Ford DJ, Pither CE, Raj PP. Electrical characteristics of peripheral nerve stimulators. Implications for nerve localization. *Regional Anaesthesia* 1984;**9**:73–7.

16. Epstein RM, Epstein RA. Electromyography in evaluation of the response to muscle relaxants. In: Katz RL, editor. *Muscle relaxants*. Amsterdam: Excerpta Medica; 1975. p. 299.
17. Szmuk P, Pavilizza E, Warters RD, Ezri T, Gebhard R. An evaluation of the T-line Tensymeter continuous noninvasive blood pressure device during induced hypotension. *Anaesthesia* 2008;**63**:307–12.
18. Moore C, Dobson A, Kinagi M, Dillon B. Comparison of blood pressure measured in the arm, ankle and calf. *Anaesthesia* 2008;**63**:1327–31.
19. Pereira E, Prys-Roberts C, Dagnino J, Anger C, Cooper GM, Hutton P. Auscultatory measurement of arterial blood pressure during anaesthesia: a reassessment of the Korotkoff sounds. *European Journal of Anaesethesia* 1985;**2**:11–20.
20. Van Bergen FH, Weatherhead DS, Treloar AE, Dobkin AB, Buckley JJ. Comparison of indirect and direct methods of measuring arterial blood pressure. *Circulation* 1954;**10**:481–90.
21. Farquhar IK. Continuous direct and indirect blood pressure measurement (Finapres) in the critically ill. *Anaesthesia* 1991;**46**:1050–3.
22. O'Brien E, Waeber B, Parati G, Staessen J, Myers MG. Blood pressure measuring devices: recommendations of the European Society of Hypertension. *Anaesthesia* 2001;**322**:531–6.
23. Awad AA, Stout RG, Ghobashy MA, Rezkanna HA, Silverman DG, Shelley KH. Analysis of the ear pulse oximeter waveform. *J Clin Monit Comput* 2006;**20**:175–84.
24. Casati A, Squicciarini G, Baciarello M, Putzu M, Salvadori A, Fanelli G. Forehead reflectance oximetry: a clinical comparison with conventional digit sensors during laporotomic and laporoscopic abdominal surgery. *J Clin Monit Comput* 2007;**21**:271–6.
25. Ralston AC, Webb RK, Runciman WB. Potential errors in pulse oximetry III. The effects of interference, dyes, dyshaemoglobins and other pigments. *Anaesthesia* 1991;**46**:291–5.
26. Howell SJ, Blogg CE, Ashby MW. Modified sensor for pulse oximetry in children. *Anaesthesia* 1993;**48**: 1083–95.
27. Langton JA, Lassey D, Hanning CD. Comparison of four pulse oximeters: effects of venous occlusion and cold induced vasoconstriction. *Br J Anaesth* 1990;**65**:245–7.
28. Talke P, Stapelfeldt C. Effect of peripheral vasoconstriction on pulse oximetry. *J Clin Monit Comput* 2006;**20**:305–9.
29. Awad AA, Haddadin AS, Tantawy H, Badr TM, Stout RG, Silverman DG, et al. The relationship between the photoplethysmographic waveform and systemic vascular resistance. *J Clin Monit Comput* 2007;**21**: 365–72.
30. Secker C, Spiers R. Accuracy of the pulse oximeter in patients with low systemic vascular resistance. *Anaesthesia* 1997; **52**:127–30.
31. Monnet X, Lamia B, Teboul J-L. Pulse oximeter as a sensor of fluid responsiveness: do we have our finger on the best solution? *Crit Care* 2005;**9**:429–30. Published online September 28. doi: 10.1186/ cc3876.
32. Stewart KG, Rowbottom SJ. Inaccuracy of pulse oximeters in patients with severe tricuspid regurgitation. *Anaesthesia* 1991; **46**:668–70.
33. Yang S, Batchelder PB, Raley DM. Effects of tissue outside of arterial blood vessels in pulse oximetry: a two dimensional pulsation. *J Clin Monit Comput* 2007;**21**: 373–9.
34. Langton JA, Hanning CD. Effect of motion artefact on pulse oximeters: evaluation of four instruments and finger probes. *Br J Anaesth* 1990;**65**: 564–70.
35. Roth JV, Braitman LE. Nasal temperature can be used as a reliable surrogate measure of core temperature. *J Clin Monit Comput* 2008;**22**:309–14.
36. Edge G, Morgan M. The Genius infrared tympanic thermometer. *Anaesthesia* 1993;**48**:604–7.
37. MacKenzie R, Asbury AJ. Clinical evaluation of liquid crystal skin thermometers. *Br J Anaesth* 1994; **72**:246–9.

Chapter 15

Physiological monitoring: gasses

Patrick T Magee

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Gas analysis during anaesthesia requires continuous monitoring of respired gasses and at times, intermittent sampling of blood gasses. The different techniques described in this chapter utilize various physical or chemical properties of the gas molecules, to detect and quantify the gas. As with all clinical measurement techniques, it is important to understand the principles on which the gas analyzers are based, so that their applications and limitations are recognized.[1]

RESPIRATORY GAS SAMPLING

In a clinical setting, respiratory gas sampling and analysis is important for a number of reasons: ensuring that the gas supply to the patient contains adequate oxygen and volatile anaesthetic agent and ensuring adequacy of ventilation by capnography, which also gives some information about the circulation.[2]

A number of factors may affect or complicate gas sampling. Common to all methods is the delay in the sample reaching the analyzer and the response time of the analyzer itself. Also, not all analyzers return the sample to the breathing system. This is advantageous when the gas analyzer alters the integrity of the gas molecule. However, if low fresh gas flows are being used in a circle breathing system, the non-return of gas samples being removed at 150 ml min^{-1} significantly increases the fresh gas requirement.

Following a step change in the gas concentration, delay in response time of the analyzer is due to two factors. The first is the *delay time* or *transit time:* the time it takes for the sample to get from the patient's airway to the gas analyzer. The second is the *response time* or *rise time* of the analyzer. The response time is usually considered to be the time taken for an analyzer to respond to within 90–95% of an actual step change in gas concentration. A step change can be produced in one of three ways: by moving a gas sampling tube rapidly into and out of a gas stream; by bursting a small balloon within a sampling volume containing a gas sample; or by switching a shutter to a gas sample volume using a solenoid valve. Fig. 15.1 shows how these are related.

Most modern analyzers use side stream sampling, where the sampling tube takes the gas sample to the analyzer. It is important that only the recommended sample tubing be used for the analyzer concerned, because various types of tubing may absorb some of the gas mixture as well as water vapour to different extents; calibration of the analyzer will have allowed for this only with the recommended tubing. Gas analyzers sample gas at a rate of between 50 and 200 ml min^{-1}. If the sampling rate is higher than this, or if the tubing is too long or too wide, the sampling waveform will be distorted, thereby reducing accuracy. The delay time depends to great extent on the sampling rate and on the length of the sampling tube, which should be as short as possible.

In trying to sample gasses at the end of expiration, it is important to sample as close to the patient's trachea as possible, particularly where the tidal volume is small and the respiratory rate is high, as when anaesthetizing infants. Some systems have a sampling catheter that extends down

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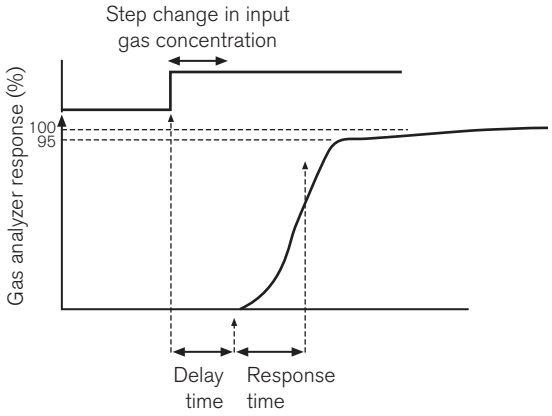


Figure 15.1 The response of gas analyzers to a step change in gas concentration.

the endotracheal tube, but this adds to airway resistance where that tube is of small diameter. Most systems, however, have a sampling port attached to the breathing system adjacent to the artificial airway. It is important that the software within the analyzer can detect minima and maxima in the respiratory waveform, and associate these with inspiratory and expiratory gas concentration values appropriately. It is still possible, however, for a gas sample, taken, for example, from the patient end of a coaxial Mapleson D breathing system, a type of T-piece, to give erroneously low end tidal readings, due to confusion between inspiratory and expiratory gas flows.

Some analyzers have used mainstream sampling instead, where the analyzer itself sits astride the artificial airway outside the mouth. This is a bulky addition to the airway but it eliminates transit time, and is reported to be more useful in detecting sleep apnoea than sidestream analyzers.[3] There is also less of a problem dealing with water vapour in the sample, and the sample is not degraded by the analyzer. An example is the Hewlett Packard infrared CO_2 analyzer, shown in Fig. 15.2. The sensor fits onto a sampling chamber inserted into the breathing system. The miniaturized unit contains a motor with filter wheel, an infrared (IR) source and detector, and IR transparent glass for the optical pathways. A second fully encapsulated optical window on the airway adapter provides a reference for calibration.

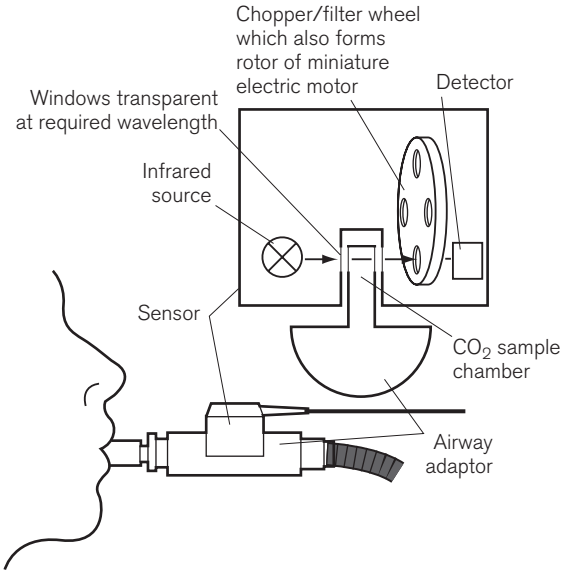


Figure 15.2 The Hewlett Packard infrared (IR) gas analyzer, using mainstream sampling.

GAS CONCENTRATION MONITORING

Refractometry

Refractometry is a technique that detects the difference in refractive indices between two media, one of which contains the gas sample of interest (e.g. air and anaesthetic vapour), the other which is of otherwise identical constitution (e.g. air only). The refractive index of a medium is a measure of the ratio of velocity of light in a vacuum to the velocity of light in that medium. The molecules of the medium delay the transit of the light wave through it; the delay depends on the number of molecules and hence on the density of the medium; thus refractive index for a gas medium depends on its concentration, pressure and temperature. When a light beam passes through parallel slits whose width is of the same order of magnitude as the wavelength of the light, interference patterns are produced by light waves arriving in phase (bright fringe) and 180° out of phase (dark fringe) with each other. When two such sets of fringes are formed from light passing through gas samples with differential velocities, they are displaced relative to each other and the displacement between them can be measured as a difference in refractive indices.

The refractometer is included, not because it is frequently used clinically as a gas analyzer, but because it is an accurate standard of gas analysis against which others are compared. All gasses and volatile agents can be quantified using this method, since all possess the physical property of refractive index. Laboratory refractometers are used by vaporizer manufacturers to calibrate vaporizers. A diagram of the portable version of the refractometer (the Riken refractometer) is shown in Fig. 15.3. Light from a small light source is collimated into a beam, which is split into two parallel beams by prisms. One beam is passed through the reference chamber containing, say, air, and the other is passed through the measuring

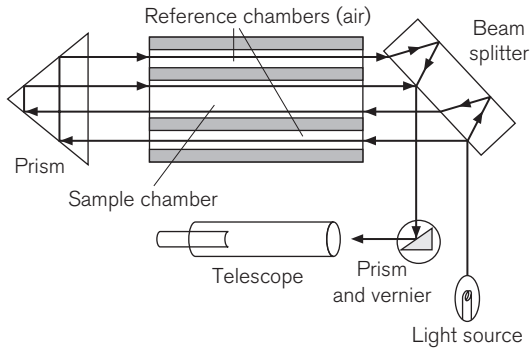


Figure 15.3 The principles of the portable refractometer.

chamber containing air and another gas/vapour species, for example sevoflurane, whose concentration is to be measured. The beams are reflected back through the chambers to enhance the effect. Each beam is passed through a slit, producing interference fringes. As discussed above, the effective path lengths of the two beams differ and two sets of fringes are produced. These can be realigned by a knob controlling a vernier scale and the amount of realignment is taken as a measure of refractive index change from the reference sample, and thus of concentration of the gas under test.

Infrared absorption spectroscopy

The interatomic bond between dissimilar atoms of a molecule absorbs radiation in the infrared (IR) range. Thus, molecules such as nitrous oxide, carbon dioxide, water vapour and the volatile agents absorb IR light, but oxygen, nitrogen and helium do not. Polyatomic molecules of different species absorb IR radiation across a range of IR wavelengths and there is frequently overlap between species. However, different polyatomic species absorb maximally at characteristic wavelengths within the IR bandwidth, which means that it is possible to identify the gas molecule as well as quantify the gas concentration. The amount of absorption (A) of any radiation by any substance is governed by the *Beer-Lambert* Law, which links A to the intensity of the incident radiation I_i, the non-absorbed, transmitted radiation I_t, the extinction coefficient ε, the path length L and the concentration C of the absorbing substance, where:

$$A = \log_{10} \frac{I_i}{I_t} = \varepsilon L.C$$

Since CO_2 analysis is important in anaesthesia, such devices are available in most anaesthetic locations in modern operating rooms. Fig. 15.4 shows the principle of the *Luft* or non-dispersive type of IR gas analyzer, which

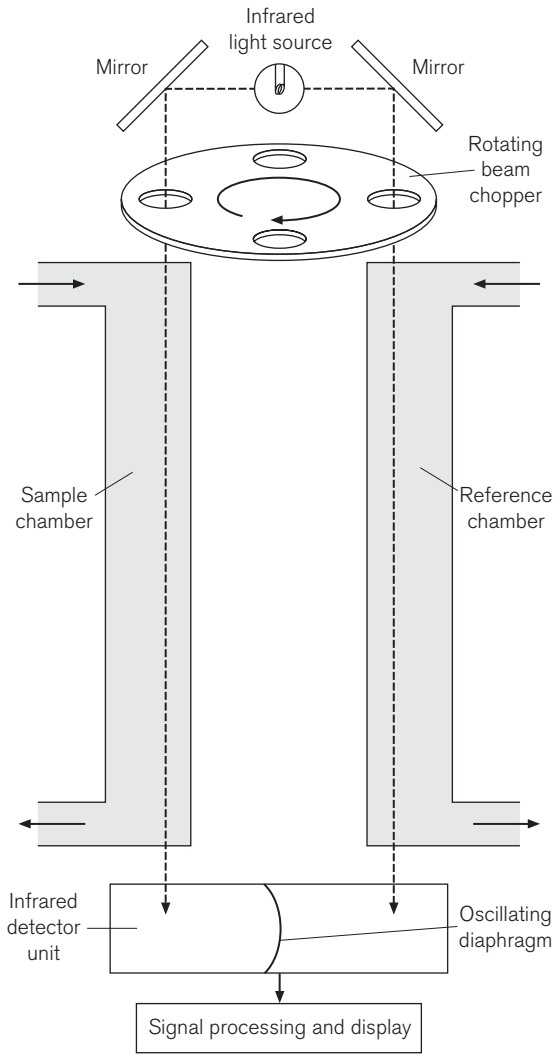


Figure 15.4 The principles of the infrared spectrophotometer.

uses a differential technique to minimize error. An IR light source emits radiation at between 1 and 15 μm. A 'chopper' wheel, with small windows containing optical filters at the rim, rotates in front of the light source at 25–100 Hz to prevent the gas sample from overheating; additionally the filters allow the passage of IR light of a narrower bandwidth than the source itself, to match the wavelength of maximum absorption of the gas under study. This window might contain a 4.26 μm filter for CO_2, a 3.3 μm filter for halothane, isoflurane and enflurane, and a 4.5 μm filter for N_2O. The light passes through this to two chambers, a reference chamber and a sample chamber; the sample

chamber contains the same background gas as the reference chamber, but additionally contains the gas under analysis. The light, which is not absorbed (the transmitted light) in each sample cell, is passed to a pair of air-filled detector chambers, separated by a diaphragm. Because different amounts of IR light are absorbed in the reference and sample chambers, different amounts of light are, therefore, transmitted to the detector chambers, heating the air in them differentially. Because the chopper wheel produces an oscillating IR source, the changing air pressure in the two detector chambers causes the diaphragm separating them to oscillate. The diaphragm usually forms half of a capacitor and is, therefore, a component of electrical circuitry, which amplifies and processes the signal, to produce an output signal, which is proportional to the gas concentration.

There are a number of sources of error in IR spectroscopy. Although CO_2, N_2O and CO absorb IR light maximally at 4.3, 4.5 and 4.7 μm respectively, there is considerable overlap in the absorption spectra of these gasses, particularly in the 3–5 μm range. There is also the phenomenon of *collision broadening*, where the presence of one gas may broaden the IR absorption spectrum of another. This effect is not only caused by IR absorbing agents themselves (such as CO_2 and N_2O) on the absorption spectra of other IR absorbing agents, but by other non-IR absorbing carrier gasses, such as helium, argon or hydrogen. It has been reported, for example, that in a gas mixture containing 79% helium in oxygen, an IR analyzer under-reads CO_2 values.[4] Desflurane,[5] cyclopropane,[6] acetone and alcohol[7] all produce errors in IR spectroscopy of respiratory gasses. There are electronic correction factors in the analyzers to allow for the presence of N_2O in the gas mix containing CO_2. Since water is a strong absorber of IR across the bandwidths of interest, it must be eliminated in the sampling process, in order to avoid some inaccuracy. If one wavelength is used for all volatile agent detection, then that agent must be selected manually, or the device will be inaccurate, if either the wrong agent is selected or if there is a mixture of agents present. Some analyzers use the 10–13 μm bandwidth to detect volatile agents, where there is less chance for interference between absorption spectra of the volatile agent and other gasses.

Infrared spectroscopy detects numbers of molecules in a gas sample and thus constitutes a partial pressure analyzer. Therefore, error can be introduced if the pressure of the gas sample changes or if there is ambient pressure change, against which the device is calibrated. If the gas sample pressure changes due to back pressure from a ventilator, or if there is significantly low pressure in a sampling tube, the partial pressure of the gas being analyzed will change, without there being a real change in the fractional concentration of the gas. Similarly, if the device is calibrated at sea level and subsequently used at altitude, there will be an error in calculating gas concentration. Therefore, calibration should be carried out frequently and under the conditions of intended use, or the device should be configured to display partial pressure rather than concentration.

The 90–95% response time of IR analyzers to a step change, including transit time and rise time, should be under 150 ms. Water vapour blocking the sampling tube sometimes causes the response time to increase. To deal with water vapour, most devices have either a water trap or use sample tubing, which absorbs water vapour. IR capnographs are accurate to about 0.1% in a range of CO_2 up to 10%.

A variant of the IR analyzer described above utilizes a combination of IR and photo-acoustic spectroscopy. The gasses under test are drawn through a small measurement chamber that is irradiated by an IR source. This IR source is first 'chopped' by a rapidly rotating wheel that has three concentric bands of holes that cause the IR energy passing through the holes to pulsate at three *different sampling rates*. Each part of the now divided beam passes through a separate narrow band optical filter allowing through only that *wavelength* that corresponds to the maximum absorption of the gas/vapour to be tested (i.e. CO_2, N_2O and volatile agent). The components of the IR light, now differentiated both by sampling frequency and wavelength, are absorbed by the gas molecules, which are caused to vibrate at different frequencies, leading to a set of audible, fluctuating pressure waves. The families of audible pressure waves are detectable by microphone. Fig. 15.5 shows the Bruel and Kjaer model of the photo-acoustic IR spectroscope. Note that, since oxygen does not absorb IR light, this device has a separate sampling channel that uses magneto-acoustic spectroscopy to measure oxygen (see the section on Paramagnetic oxygen analyzers).

The signal from all gaseous components is taken from the microphone and filtered electronically into the four components of O_2, CO_2, N_2O and volatile agent. Separation and, therefore, gas identification occurs by audio frequency identification and quantification of the gas occurs by audio-amplitude. Note that the device does not distinguish between different volatile agents. The advantages of photo-acoustic IR spectroscopy over conventional IR spectroscopy include stability, zero drift, reduced need for calibration over prolonged periods and fast response time.

A more recent design of infrared absorption spectroscopy does away with the moving parts of the chopper wheel. Light from a broad-spectrum infrared source passes through a cuvette containing sampled gas. The Dräger ILCA (Fig. 15.6) then uses a prism as a beam splitter. On the four sides of the multichannel detection chamber are appropriate infrared filters and sensors for the maximal absorption wavelengths of each gas. By using two such modules in series the *gas bench* can quantify CO_2, N_2O and the individual volatile agents in the gas mixture.

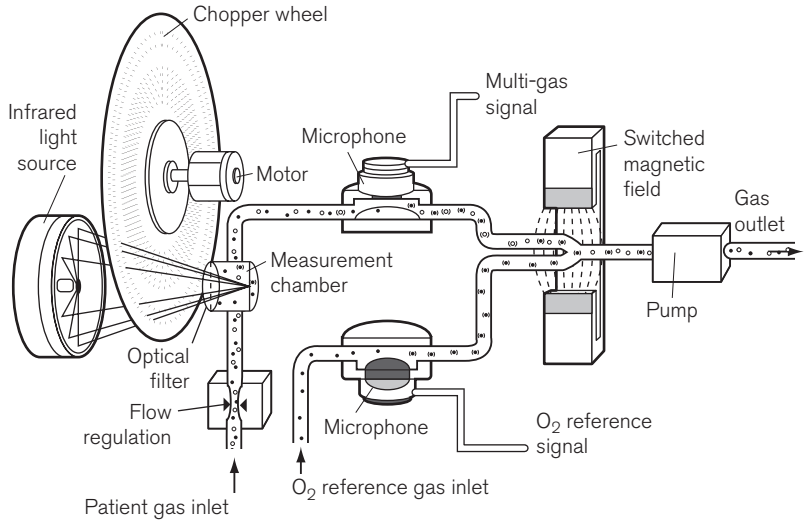


Figure 15.5 The components of the Bruel and Kjaer photo-acoustic and magneto-acoustic spectrophotometer.

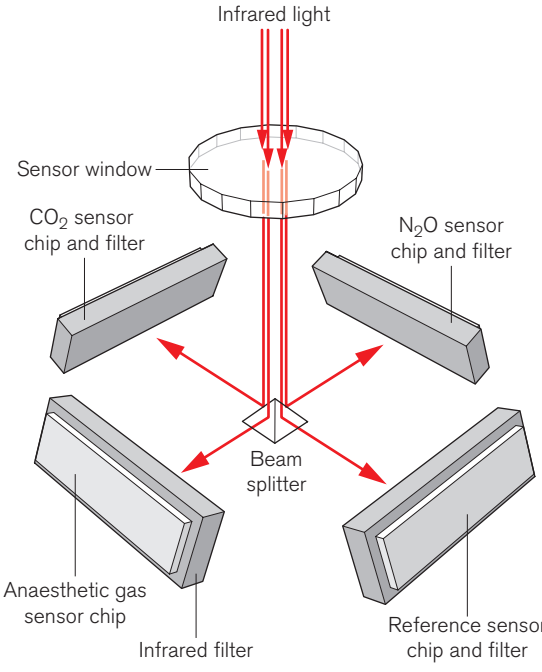


Figure 15.6 Working principles of Dräger ILCA gas analyzer.

Mass spectrometry

Mass spectrometry identifies gas molecules by a two-stage process; the molecules are first bombarded in a vacuum with electrons, which converts them into charged particles; the charged particles are then separated in a magnetic field according to the ratio of their mass to charge. The method can, therefore, identify and quantify all molecular species, providing there are no unexpected components in a gas mixture, such as the propellant in a bronchodilator or acetone in the expired breath of a diabetic. It is still considered to be the standard against which other gas monitoring techniques are compared, and is more frequently used as a laboratory technique than a clinical one. However, it was available for use as a time-shared, multi-site sampling device before the advent of cheaper IR spectroscopy. One example of a mass spectrometer is shown in Fig. 15.7A. Essentially it has three stages: the first stage is where the sampled gas is drawn into a low-pressure chamber. In the second stage, which is the main part of the device, the sample is drawn into an ionization chamber by an even lower pressure (about 1 mmHg), where the gas molecules are bombarded with electrons. Finally, they are drawn into a dispersion chamber, where they are influenced by a magnetic field. This third stage is where the deflected and separated beams of charged molecules are detected and the signal is processed and displayed.

In the vacuum chamber of the second stage of the device, the bombardment of the molecules, by a transverse beam of electrons, results in ionization of the molecule, usually to a single positive charge. The ions of the different molecular species then accelerate towards an electrically negatively charged plate, the acceleration plate, and out through a small hole, *the molecular leak*, into the dispersion chamber. From here the path of the ions is influenced by a magnetic field. The ions are deflected according to mass; the lightest being deflected the most. The different species of gas are thus separated according to their mass: charge ratio. In the third stage, the ions reach the photovoltaic

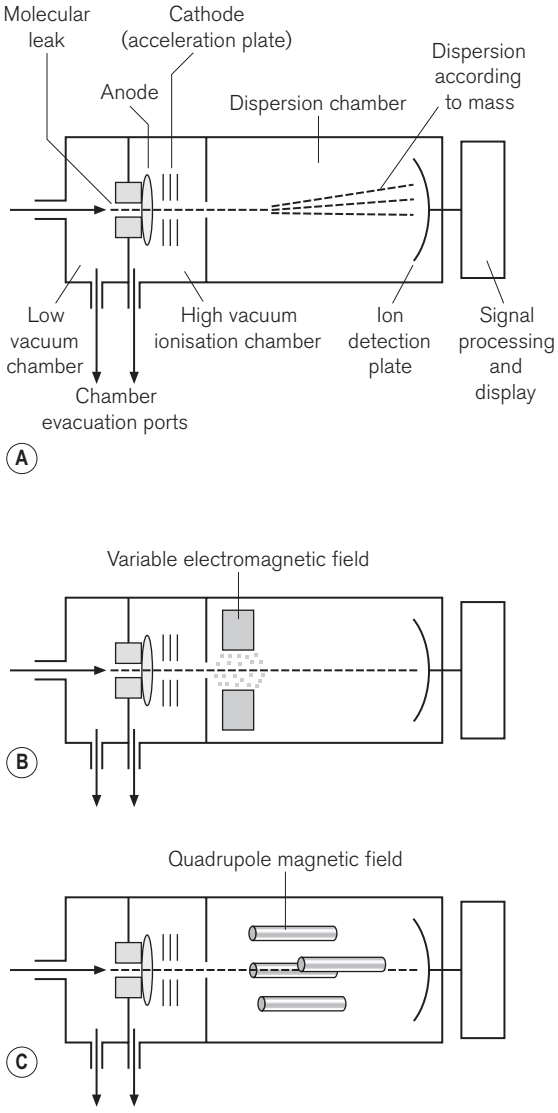


Figure 15.7 **A.** Principles of a mass spectrometer. **B.** Components of a magnetic sector mass spectrometer. **C.** Components of the quadrupole mass spectrometer.

detectors, where the rate of arrival of the ions is proportional to the partial pressure of the gas. The signal is processed, amplified and displayed.

Two different types of mass spectrometer exist, depending on the way in which the magnetic field is produced in the second stage and the mechanism of ion detection in the third stage.

In the *magnetic sector mass spectrometer* (Fig. 15.7B) the magnetic field lies at right angles to the path of ions and is produced by a combination of electrical and fixed magnetic fields. By varying the voltage on the acceleration plate, the velocity of the ions entering the magnetic field can be changed. This allows the deflected ion beam to be directed across a single detector plate and different components to be detected in turn. It is, therefore, possible to separate components of the gas sample according to their mass : charge ratio.

More commonly seen these days, is the *quadrupole mass spectrometer*. This device is more compact and allows better discrimination between the ionic components from a gas mixture. The magnetic field is produced by a combination of DC electrical and AC radio frequency fields. As shown in Fig. 15.7C, the quadrupoles consist of four rods with opposite pairs electrically connected. By careful tuning of the radio frequency component of the magnetic field, only ions of a given mass : charge ratio proceed through the quadrupole to the detector, all other ions oscillating and colliding with the device. By a combination of changing the voltage on the acceleration plate and of judiciously tuning the magnetic field, a spectrum of mass : charge components can be detected and quantified. By scanning at 50 Hz, it is possible to produce a continuous record of gas concentrations.

The respiratory mass spectrometer is accurate, giving good gas identification and quantification, requiring only 20 ml min^{-1} gas sampling rate, with a 100 ms response time. However, it does have some disadvantages. The second stage of the device operates under almost vacuum conditions; this requires a high-quality, continuously running pump. If the device itself is at some distance from the sampling site, as it used to be in the days of time-sharing of a single device, significant delay time may be added to the response time. Water condensation can be avoided by heating the sampling tube, but the response time for water vapour may still be longer than for other components.

Some molecule types may lose two electrons rather than one in the ionization process, and, therefore, become doubly charged ions rather than single. They then behave within the magnetic field like an ion with half the mass, which leads to confusion in interpretation. Furthermore, the ionization process can lead to fragmentation of a molecule, so that a mass spectrum appears at the output rather than a single peak. However, this anomaly is useful to distinguish the gas components with the same molecular mass, which would otherwise be difficult to distinguish. These include N_2O and CO_2 (44 Daltons [Da]), or N_2 and CO (28 Da). N_2O is fragmented into NO, O_2, N_2, N and O, while CO_2 is fragmented into O_2, C_2, C and O. Rather than try to detect and distinguish both gasses at 44 Da, N_2O can be detected at a subordinate peak of 30 Da and CO_2 at 12 Da. Because the fragmentation is predictable, the amplitude of the subsidiary peak can be used as a measure of the parent peak and, therefore, of gas concentration. Using appropriate low pass and high pass filters, obfuscating peaks in the spectra can be removed.

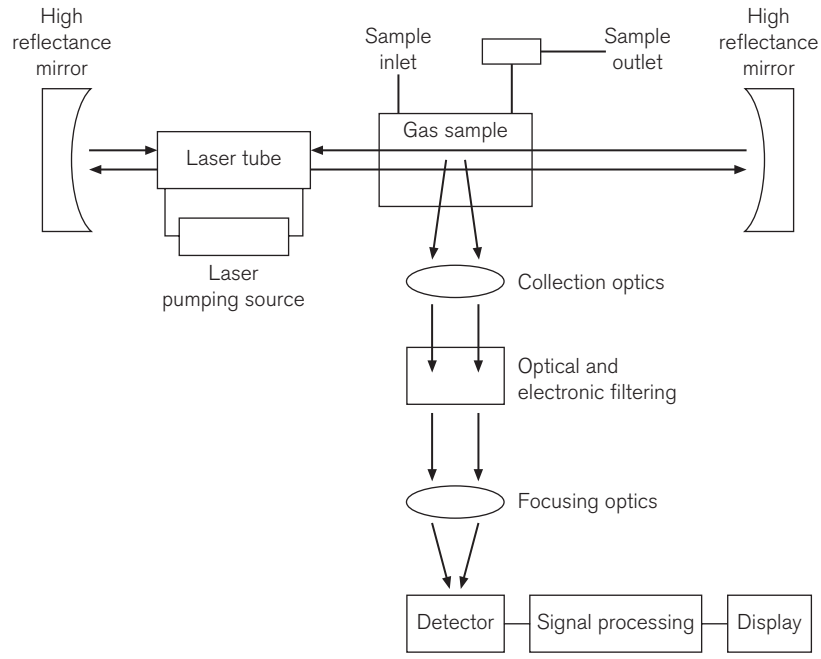


Figure 15.8 The components of the Raman spectroscope.

Raman spectroscopy

Rayleigh scattering refers to the scattering of light by particles in its path of size up to one-tenth the wavelength of the light and occurs *without* any loss of energy or change of wavelength. Being wavelength dependent, this phenomenon gives us nature's blue sky because of the increased scattering of blue light. However, a small fraction of the incident light, about 10^{-6}, is scattered *with* a loss of energy and a change of wavelength characteristic of the molecule off which the light is being reflected; this is *Raman scattering*. Raman spectroscopy has been used in industry for years as a means of identifying solids, liquids and gasses, but has had to await the advent of powerful laser light sources and sensitive photocell detectors to be useful in a clinical setting for breath-by-breath analysis.

Fig. 15.8 shows a diagram of a Raman spectroscope, incorporating an Argon laser source of wavelength 485 nm, high reflectance mirrors to concentrate the laser beam, a gas sampling chamber, appropriate optics, a detection system, and microprocessor and display system. Table 15.1 shows characteristic wavelength or frequency changes, which identify different gasses. If plotted graphically, the amplitudes of the frequency shifted peaks are proportional to the gas concentrations.

In Raman spectroscopy, each gas is independently analyzed, including CO_2, N_2O, volatile agents, O_2, N_2 and water vapour. This is done by first filtering the Argon laser light at 485 nm, then passing the light through a system of optics to collect and focus the scattered low level light, then through an electronic filtering system; the very low levels of light are then detected by a photomultiplier tube. The response time is 100 ms and the device is, therefore, capable of breath-by-breath analysis. The sample is not altered by the process and can, therefore, be returned to the breathing system, which is a significant advantage in low flow anaesthesia,[8] although there is some overlap between gas species.[9] However, the devices require a great deal of electrical power and are noisy.

Table 15.1 Frequency shifts in different gasses detectable by Raman spectroscopy (N_2O and CO_2 give two characteristic frequency shifts).

GAS	FREQUENCY SHIFT WAVE NUMBER (cm^{-1})
Isoflurane	995
Halothane	717
Enflurane	817
Nitrous oxide	1285, 2224
Carbon dioxide	1285, 1388
Oxygen	1555
Nitrogen	2331
Water	3650

The piezoelectric (Engstrom Emma) gas analyzer

A pair of piezoelectric crystals connected to an electrical power source is made to resonate, with a characteristic frequency difference. The frequency difference occurs because one of the crystals is coated in silicone oil, which absorbs a volatile anaesthetic agent to which it is exposed. This changes the natural frequency of the crystal, and the frequency difference between the crystals changes by an amount dependent on the amount of volatile agent absorbed. Manual identification of the agent is required, but the device is remarkably accurate and stable.[10]

The paramagnetic gas analyzer

Most gas molecules are repelled by a magnetic field and are, therefore, termed diamagnetic. Two gasses, oxygen and nitric oxide, are attracted into the field and are termed paramagnetic. This property enables oxygen concentrations to be analyzed and is due to the presence of unpaired electrons in the outer shell of an oxygen molecule, which is able to generate force in a magnetic field. A paramagnetic oxygen analyzer of the original type is shown in Fig. 15.9A. Two glass spheres, suspended between the poles of a magnetic field, are filled with nitrogen, a weakly diamagnetic gas. The glass spheres are arranged in a dumbbell shape, suspended by a thread, tensioned to keep the dumb-bell in the plane of the magnetic field. Zeroing should be carried out in the carrier gas destined to have oxygen added to it at a later stage. When a gas mixture containing oxygen is drawn through the analyzer, oxygen is attracted into the magnetic field, displacing the nitrogen filled spheres away from it. The detection system can either be a deflection measurement (Fig. 15.9A) or a null deflection type (Fig. 15.9B), which measures the current required in the circuitry to restore the pointer to its null point. These devices are accurate to within 0.1% O_2, but are adversely affected by pressurization, vibration, water vapour and high flow rates; there is also a slow response time of up to 1 min.

An analyzer has been developed to overcome these disadvantages (Fig. 15.10). A sample of gas to be measured is drawn continuously through one of two capillary tubes at the same rate as a reference gas sample (usually air) is drawn through the other. A powerful electromagnet,

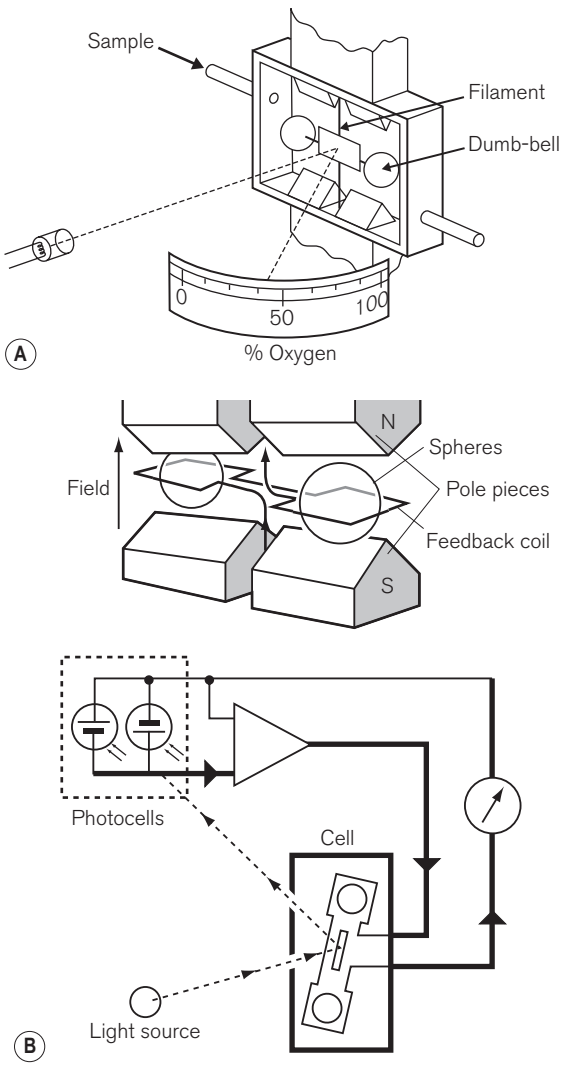


Figure 15.9 **A.** A simple paramagnetic oxygen analyzer. **B.** A null deflection paramagnetic oxygen analyzer. *With permission from Kenny GNC, Davis PDD (2003)* Basic Physics and Measurement in Anaesthesia, *5th edn. London: Butterworth Heinemann.*

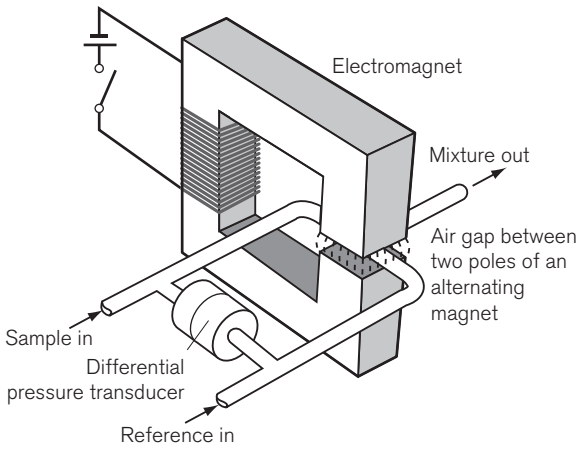


Figure 15.10 A modern paramagnetic oxygen analyzer.

producing a pulsed (on/off) magnetic field is placed over the junction of the two tubes. The magnetic field, when switched on, attracts the oxygen from both tubes in proportion to their concentrations. This causes an intermittent differential reduction in pressure upstream in the tubes that is detected and measured by a pressure transducer, and can be calibrated so it displays oxygen concentration. This is the basis of the Datex paramagnetic oxygen analyzer.

The pulsed magnetic field may be replaced by an alternating one that has a frequency of 110 Hz. This produces differential oscillating pressures of 20–50 μbar in the capillary tubes. The oscillations are transduced into a sound signal, the amplitude of which is directly proportional to the O_2 concentration in the sample. This is magneto-acoustic spectroscopy and forms the oxygen analyzing device in the Bruel and Kjaer gas analyzer.

It has been reported that a gas mixture containing desflurane interferes with the accuracy of a paramagnetic oxygen analyzer.[5]

Dräger PATO

The paramagnetic property of oxygen can be exploited in different ways as the basis for oxygen concentration measurement. In the Dräger PATO O_2 sensor a very small measurement chamber lying between two large magnets has heating and temperature measuring elements on either side. The presence of oxygen in the gas mixture reduces the thermal conductivity of the gas between the magnets producing a rapid response oxygen analyzer (Fig. 15.11A and B).

Fuel cells and polarographic cells

These techniques are used for analyzing oxygen and are included together, since they are similar electrochemical techniques.

The Clarke polarographic electrode is shown in Fig. 15.12A. It consists of a cellophane covered platinum cathode and Ag/AgCl anode in a phosphate and KCl electrolyte, between which a potential difference of −0.6 V is applied by the battery as shown. At the cathode the following reaction takes place:

$$O_2 + 2H_2O + 4e^- \rightarrow 4OH^-$$

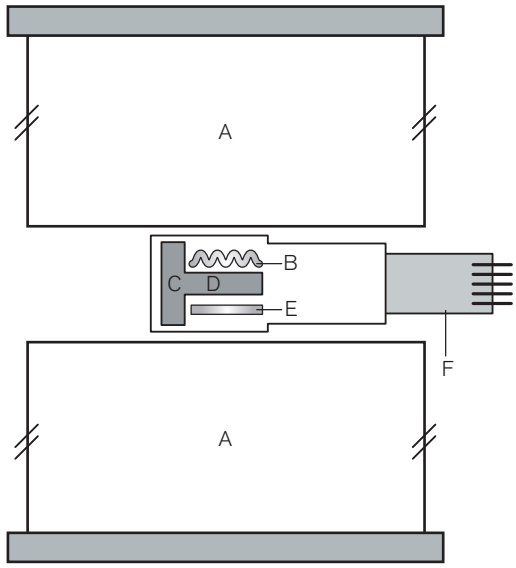
Electrons are provided by the cathode, to be consumed by the O_2 in the gas or blood sample. At the anode the following reactions take place:

$$4Ag \rightarrow 4Ag^+ + 4e^-$$

$$4Ag^+ + 4Cl^- \rightarrow 4AgCl$$

The rate of diffusion of oxygen into the electrolyte solution is a rate-limited process, generating a current proportional to the pO_2 of the gas sample.

The problem with the polarographic electrode is the dependence on a battery and the reduction of any N_2O present at the cathode when it is contaminated by Ag^+ ions. A fuel cell is a similar device, which consists of a gold cathode and a lead anode (Fig. 15.12B). The same reaction occurs at the cathode as in the polarographic electrode. A polarizing voltage is generated by the process and 'fuel'



A = Electromagnet
B = Heating element
C = Gas pathway
D = Measurement chamber
E = Sensing element
F = Electrical connector



Figure 15.11 **A.** Schematic of Dräger Pato paramagnetic oxygen analyzer. **B.** Dräger PATO O_2 sensor complete, and disassembled to show magnets and interposing sensor strip carrying gas pathway and measurement compartment (the diameter of the cylinders is about 5 cm).

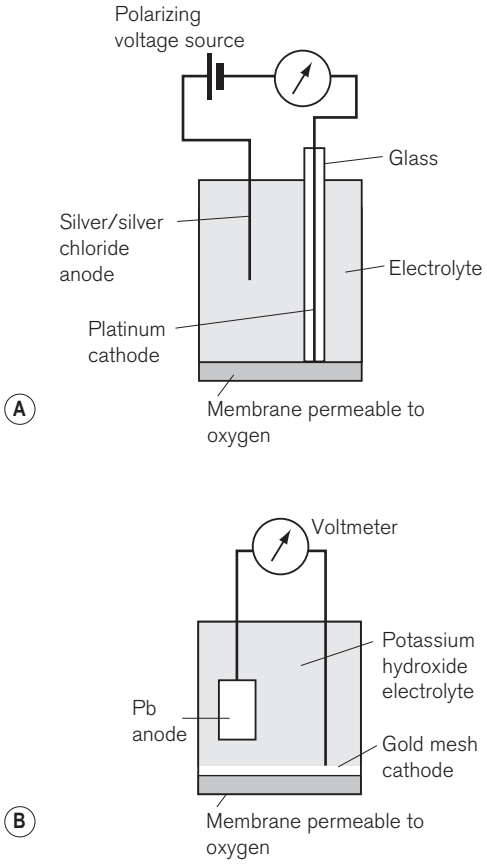


Figure 15.12 **A.** The Clarke polarographic electrode. **B.** The Galvanic fuel cell.

is consumed, such that the cell needs to be replaced at intervals of 6 months to a year. The fuel cell's response time is slow, making them acceptable, but less suitable for breath-by-breath analysis than other methods.[11]

Nitric oxide measurement

Nitric oxide (NO) is a pulmonary vasodilator and as such, has found its place in an intensive care environment. Since NO causes methaemoglobinaemia, it is important to be able to measure its concentration. Since nitrogen dioxide (NO_2) is a degradation product of NO and causes pulmonary oedema, it too must be detectable.

All industrial analyzers developed for NO analysis are designed for use in environmental control and, therefore, for gas mixtures containing 21% or less of oxygen. Medical gas mixtures frequently contain more oxygen than this and the accuracy of devices available is not guaranteed. The two methods available are the *chemoluminescent* and the *electrochemical* methods.[12]

The chemoluminescence analyzer is based on the reaction of NO with ozone (O_3) in a chamber to produce energized NO_2 molecules, which emit photons of light in the range 590–2600 nm. The amount of light emitted is proportional to the original concentration of NO. In a parallel system there is a catalytic converter before the reaction chamber to reduce NO_2 in the sample to NO. The total oxides of nitrogen are then measured as NO and the difference in the two readings is the NO_2 concentration.

The electrochemical analyzer generates a potential difference proportional to the NO_2 concentration. This is due to the migration of ions generated from the reaction between NO or NO_2 and the electrolyte. A separate cell is used for NO and NO_2.

MEASUREMENT OF RESPIRATORY VOLUMES

Measurement of respiratory volumes is carried out by one of a number of techniques, most of which involve integration of flow measurement. These include hot wire anemometry, ultrasonic detection of flow vortices, and venturi flow measurement. Devices for volume measurement, which can be considered not to be flow integration techniques, include a turbine respirometer (Wright's) and the positive displacement volume meter (Dräger). The principles of these techniques are discussed in Chapter 2.

BLOOD GAS ANALYSIS

Although it is now possible to continuously measure oxygen saturation with pulse oximetry; and mixed venous oxygen saturation, and even pO_2, pCO_2 and pH with intravascular electrodes, it is still much more common to measure these variables by *in vitro* blood gas analysis.

The blood gas analyzer needs a very small, heparinized blood sample, 100–300 µl. The blood is passed through four electrodes sequentially in a temperature stabilized cuvette, shown in Fig. 15.13. The electrodes measure the partial pressure of dissolved oxygen using a Clarke electrode, carbon dioxide using a Severinghaus electrode and pH with a conventional glass electrode; the fourth electrode is a reference electrode. Such a blood gas analyzer frequently measures Hb and some biochemistry as well.[12] Derived values from such a device include O_2 saturation, O_2 content, bicarbonate, base excess and total CO_2.

The Clarke polarographic electrode for measuring pO_2 has been described in the section above on gas analyzers and is suitable for both respiratory and blood O_2 analysis. The membrane is designed to allow only oxygen to cross it, rather than blood.

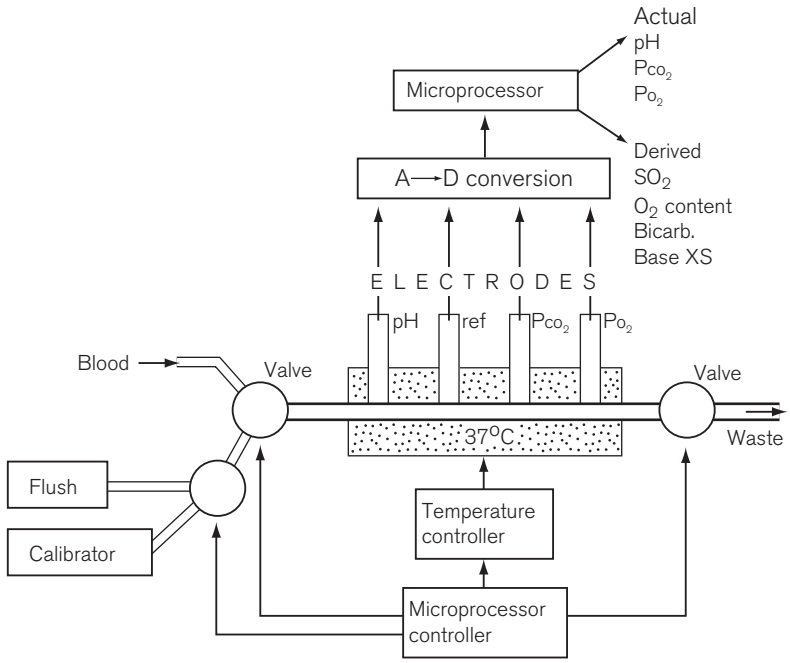


Figure 15.13 The components of a blood gas analyzer.

pH electrode

A *pH unit*, which is the measure of hydrogen ion activity in a liquid, is defined as:

$$\mathrm{pH} = \log_{10} \frac{1}{[\mathrm{H}^+]}$$

In words, this can be described as 'the negative logarithm, to base ten, of the hydrogen ion concentration'.

The physical principle, on which the pH electrode is based, depends on the fact that when a membrane separates two solutions of different $[H^+]$, a potential difference exists across the membrane. In a pH electrode, the membrane is made of glass and the development of a potential difference between the two solutions is thought to be due to the migration of H^+ into the glass matrix. If one solution consists of a standard $[H^+]$, the pH of the other solution can be estimated by measurement of the potential difference between them. The glass membrane used is selectively permeable to H^+. No current flows in this device, which therefore does not wear out, in contrast to the Clarke electrode, in which current does flow and which does need periodic replacement.

The pH measurement system is shown diagrammatically in Fig. 15.14 and consists of two half electrochemical cells. It has in one half an Ag/AgCl electrode and in the other a $Hg/HgCl_2$ (calomel) electrode. Each electrode maintains a fixed electrical potential. The Ag/AgCl electrode is surrounded by a buffer solution of known pH,

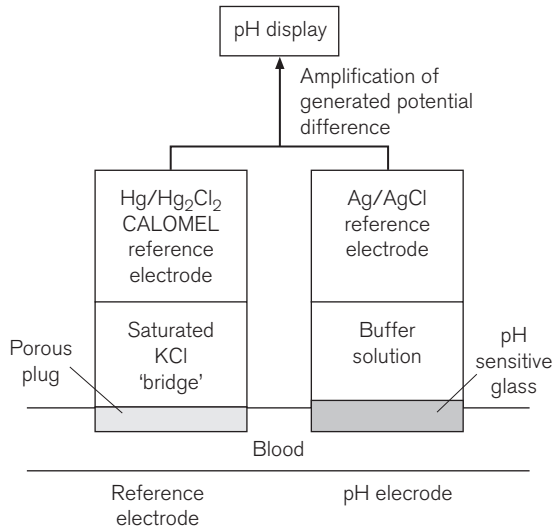


Figure 15.14 The pH electrode.

surrounded by the pH sensitive glass. Outside the glass membrane is the test solution, usually blood, whose pH is to be measured. It is the potential difference across the glass, between these two solutions, which is variable. The blood or other solution is separated from the calomel electrode by a porous plug and a potassium chloride

salt bridge to minimize KCl diffusion. The potential difference across the system is about 60 mV per unit of pH change at 37°C. The internal electrical resistance is high and in order to maximize the device's accuracy, a voltmeter (or other measuring device) of very high internal resistance must be used to minimize current drawn from the system.

The pH electrode actually responds to hydrogen ion activity rather than hydrogen ion concentration. These two variables coincide at infinite dilution of solution, but might otherwise differ, because of molecular interaction between ionic species. The electrode is, therefore, calibrated against standard buffer solutions of pH 6.841 and 7.383. The glass of which the pH sensitive electrode is made consists of 72% SiO_2, 22% Na_2O and 6% CaO.

The Severinghaus pCO_2 electrode

There is a dynamic equilibrium between H^+ and CO_2. The *Henderson-Hasselbalch* equation describes this relationship:

$$pH = pK_a + \log_{10}\frac{[HCO_3^-]}{[H_2CO_3]}$$

$$= pK_a + \frac{\log_{10}[HCO_3^-]}{\alpha pCO_2}$$

where α is the Ostwald solubility coefficient, 0.003 mmol L^{-1} mm Hg^{-1} at 37°C. pK_a, the association constant of the chemical reaction concerned is 6.1. Both α and pK_a vary with pH and thus lend the calculation some inaccuracy. It was Astrup who noticed that there was a linear relationship between pH and pCO_2.

The pCO_2 electrode is essentially a pH electrode with a difference. The sensitive glass membrane is covered with another membrane, which is selectively permeable to CO_2. In the original design, a layer of water was trapped between the two membranes, allowing the reaction described above to occur. A later modification had salt and bicarbonate solution in this space (Fig. 15.15). A pCO_2 electrode, therefore, allows a pCO_2 change to generate a pH change, which is measured by the electrode.

Derived variables from a blood gas machine

Buffer base is the sum of all bases in the blood capable of buffering pH changes. These include HCO_3^-, Hb, PO_4^{3-} and the anionic parts of protein. A respiratory driven change in status, marked by, for example, the addition of CO_2, results in the formation of carbonic acid, which produces H^+ and HCO_3^-. The hydrogen ions are buffered by the total buffer base, namely: $HCO_3^- + Hb^- + PO_4^{3-} + Protein^-$.

The buffer base can be considered to be a limitless source of buffering capacity, unless the haemoglobin component is significantly reduced. The base excess, an indication of metabolic rather than respiratory acid–base disturbance, is defined as the amount of titratable acid required to correct a blood sample to a pH of 7.40 at a pCO_2 of 5.3 kPa and at 37°C.

Standard bicarbonate is the bicarbonate in a fully oxygenated blood sample at a pCO_2 of 5.3 kPa and at 37°C.

Errors in blood gas measurement include:

- air bubbles in the sample
- excess heparin
- gas storage in container plastic walls
- failure to store samples at low temperature
- metabolism in the sample
- signal processing errors
- at the O_2 electrode:
  - O_2 consumption in a Clark electrode
  - material used in electrode membrane
  - polarizing voltage
- at the pH electrode:
  - protein deposition on the electrode.

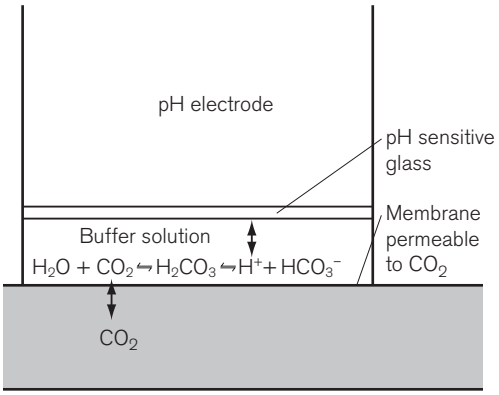


Figure 15.15 A pCO_2 electrode.

Temperature and blood gas analysis

Blood gas analyzers measure blood gas variables at 37°C. A fall in body temperature means that CO_2 is more soluble in blood and that CO_2 production slows, leading to a fall in pCO_2. This shifts the O_2 dissociation curve to the left, increasing the O_2 content. However, O_2 solubility also rises, decreasing pO_2 and measured O_2 content. pH increases with a fall in temperature. Hypothermia itself does not alter the blood gas values at 37°C and arterial blood taken at lower temperatures should be corrected to 37°C before clinical decisions are made on the results.[13]

Other ion-selective electrodes

By carefully selecting the proportions of constituents in the glass, glass ion-selective electrodes can be constructed for detection of other ions. For example, a glass electrode for measuring Na^+ concentrations consists of 71% SiO_2,

18% Al_2O_3 and 11% Na_2O. The K^+-sensitive glass electrode consists of 69% SiO_2, 4% Al_2O_3 and 27% Na_2O.

Transcutaneous blood gas analyzers

pO_2 can be measured transcutaneously by applying a Clarke polarographic electrode to the skin. If the skin is 'arterialized' by heating it, a reasonable value for pO_2 can be obtained. A modified pCO_2 electrode is used to measure transcutaneous pCO_2. A common O_2/CO_2 permeable membrane and electrolyte solution is used, ensuring a common pH for both measurements. Actual measured values do not correlate well with arterial values and should not replace formal respiratory gas analysis,[14] but trends are clinically useful and these devices are used particularly in neonatology and when patients are ventilated outside the intensive care setting where arterial blood gas sampling is problematic. Care must be taken to avoid burns due to the heated electrodes.

Intravascular blood gas analyzers

These are microminiaturized versions of the devices described above.

The co-oximeter

The co-oximeter is usually considered alongside the blood gas analyzer. The co-oximeter measures the fractional oxygen saturation of haemoglobin, taking account of the numerous different species of haemoglobin present in the blood, such as reduced Hb (HHb), oxygenated Hb (HbO), methaemaoglobin (MetHb), sulphaemoglobin and caorboxyhaemoglobin (COHb). The co-oximeter is a spectroscope, which, therefore, needs four or five different wavelengths to distinguish these different species. The Beer-Lambert law relating to light absorption is more closely adhered to than the pulse oximeter, which measures functional oxygen saturation, using only two wavelengths of light, and assumes the sample only contains reduced and oxygenated haemoglobin.

A small sample, 200–400 µl, of heparinized blood is injected into a cuvette of known path length. The red blood cells are disrupted by ultrasound or a soap solution, releasing free haemoglobin. Conventional spectroscopy then takes place on the sample at up to 17 wavelengths, which are produced either by a filter wheel with multiple narrow band filters, or by scanning with a diffraction grating or a prism. Fig. 15.16 shows the

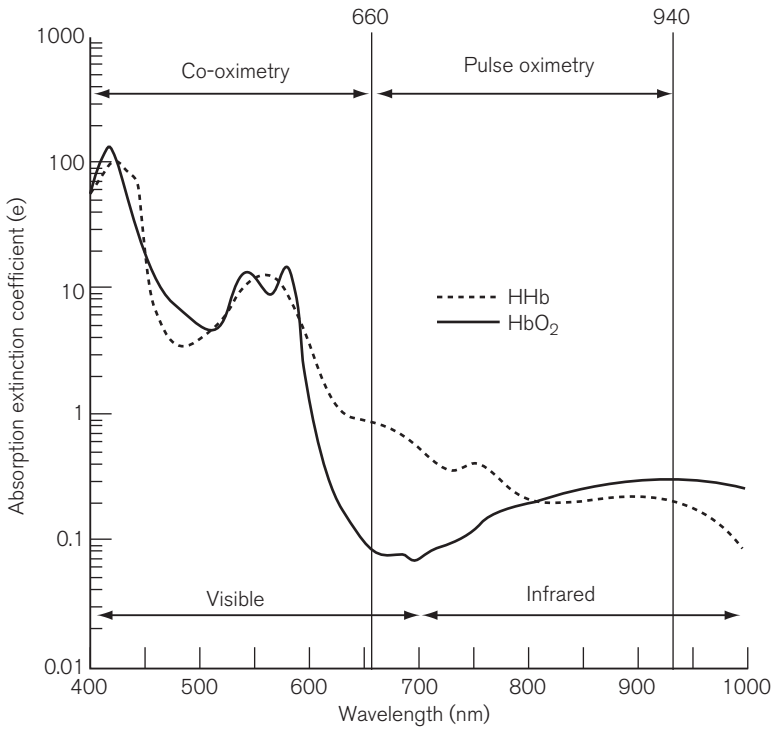


Figure 15.16 Absorption spectra for reduced and oxygenated haemoglobin showing the range of operation of a co-oximeter and of a pulse oximeter.

spectrum covered by a co-oximeter as well as that covered by a pulse oximeter. The result is usually available within 30 s and includes SO_2 %, total haemoglobin (Hb) concentration, oxygen content, methaemoglobin (MetHb) %, sulphaemoglobin %, carboxyhaemoglobin (COHb) %. Some devices are capable of distinguishing foetal haemoglobin (HbF). As Fig. 15.16 shows, the pulse oximeter and the co-oximeter operate in different spectra and, therefore, strictly speaking, SO_2 and SpO_2 should not be compared.

REFERENCES

1. Langton JA, Hutton A. Respiratory gas analysis. *BJA Continuing Education* 2009;**9**:19–23.
2. Eipe N, Tarshis J. A system of classification for the clinical applications of capnography. *J Clin Monit Comput* 2007;**21**: 341–4.
3. Yamamori S, Takasaki Y, Ozaki M, Iseki H. A flow-through capnometer for obstructive sleep apnoea. *J Clin Monit Comput* 2008;**22**:209–20.
4. Ball JAS, Grounds RM. Calibration of three capnographs for use with helium and oxygen mixtures. *Anaesthesia* 2003;**58**:156–60.
5. Scheeren TWC, Krossa M, Meriläinen P, Arndt JO. Error in measurement of oxygen and carbon dioxide concentrations by the Deltatec II metabolic monitor in the presence of desflurane. *Br J Anaesth* 1998;**80**:521–4.
6. Mason DG, Lloyd-Thomas AR. Cyclopropane and the Datex Capnomac. *Anaesthesia* 1991;**46**:398–9.
7. Foley MA Wood PR, Peel WJ, Jones GM, Lawler PG. The effect of exhaled alcohol on the performance of the Datex Capnomac. *Anaesthesia* 1990;**45**:232–4.
8. Lockwood GG, London MJ, Chakrabarti MK, Whitwam JG. The Ohmeda Rascal II. A new gas analyser for anaesthetic use. *Anaesthesia* 1994;**49**:44–53.
9. Lawson D, Samanta S, Magee PT, Gregonis DE. Gas monitoring in the OR: stability and long term durability of Raman spectroscopy. *J Clin Monit* 1993;**9**:241–51.
10. Humphrey SJE, Luff NP, White DC. Evaluation of the Lamtec anaesthetic agent monitor. *Anaesthesia* 1991;**46**:478–81.
11. Roe PG, Tyler CKG, Tennant R, Barnes PK. Oxygen analysers. Evaluation of five models. *Anaesthesia* 1987;**42**:175–81.
12. Etches PC, Harris ML, McLinley R, Finer NM. Clinical monitoring of inhaled nitric oxide: comparison chemoluminescent and electrochemical sensors. *Biomed Instrum Technol* 1995;**29**:131–40.
13. Rupp SM, Severinghaus J. Hypothermia. In: Mipler R, editor. *Anesthesia*. 2nd ed. New York: Churchill Livingstone; 1986. p. 2000.
14. Bolliger D, Steiner LA, Kasper J, Aziz OA, Filipovic M, Seeberger MD. The accuracy of non-invasive carbon dioxide monitoring: a clinical evaluation of two transcutaneous systems. *Anaesthesia* 2007;**62**:394–9.

FURTHER READING

Sykes MK. The determination of pH. In: Scurr C, Feldman S, editors. *Scientific foundations, anaesthesia*. 3rd ed. London: Heinemann; 1982. p. 108–14.

Hutton P, Hahn C, Clutton-Brock TH. Gas and vapour analysis. In: Hutton P, Prys-Roberts C, editors. *Monitoring in anaesthesia and intensive care*. London: WB Saunders; 1994. p. 194–203.

Chapter 16

Cardiac output monitoring

Robert Sekun Kong and Luigi Vetrugno

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Many anaesthetic drugs adversely affect the cardiovascular system to some degree. This response is often exacerbated by pre-existing medical or surgical conditions. Failure to take account of these factors may lead to irreversible organ dysfunction in the perioperative period. Monitoring the efficiency of the cardiovascular system and taking therapeutic steps to improve its function has become an essential part of high-quality anaesthesia. In the past, non-invasive measurements of variables such as blood pressure and pulse, gave limited information as to the state of the cardiovascular system. However, it is now recognized that inferences of cardiac output (CO) made from these measurements can be wildly inaccurate and insufficient for therapeutic intervention in the perioperative period or during critical illness.

Newer methods of assessing and optimizing cardiovascular function have been developed that allow vastly improved therapeutic intervention by anaesthetists in the perioperative period. Today, at least a dozen devices are available to measure or estimate CO quite accurately. Some of the more popular ones, based on their market presence or adoption in clinical practice, will be described in this chapter. Table 16.1 is a more comprehensive list of the available equipment and their associated technologies.

PULMONARY ARTERY CATHETER

The advent of the Swan-Ganz pulmonary artery catheter (PAC) in 1970 enabled clinicians to measure cardiac output (CO) at the bedside.[1]

For most of the last 40 years, the PAC has been the preferred tool for haemodynamic monitoring in critically ill patients. As well as CO, the PAC can measure right heart and pulmonary artery pressures, estimate cardiac preload, and evaluate global oxygen delivery and consumption. Although such a comprehensive assessment of the circulation would be assumed to enhance patient care, there is little evidence to support this notion. Several studies have failed to show that using a PAC improved the outcome of critically ill patients in general[2,3] or those with congestive cardiac failure[4] or acute respiratory distress syndrome.[5] Now and consequent to the emergence of the other less invasive technologies for measuring CO discussed later in this chapter, the last ten years has seen a continuing and dramatic decline in PAC usage.

The original PAC was a double lumen balloon-tipped catheter. A wide range of PACs is now available with one manufacturer producing around 30 different models. The PAC for adult use is typically a multi-lumen catheter, 110 cm in length and 7.0 or 7.5 F (French gauge) in external diameter. It is inserted via an introducer sheath (8.0–9.0 F depending on catheter model) into the internal jugular, subclavian or femoral vein. The proximal ports on the catheter are intended for administration of fluids or drugs, with their lumens appearing 19–30 cm from the distal tip of the catheter. To position the PAC, the balloon

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Table 16.1 A representative sample of the variety of devices available for measuring CO and their technological bases.

MONITOR	ACCESS REQUIRED	INVASIVENESS	CO-MEASUREMENT METHOD	CO-UPDATE INTERVAL	MANUFACTURER AND WEBSITE
Vigilance II	Catheter in pulmonary artery	High	Thermodilution – thermal filament	<5 min	Edwards Lifesciences, USA www.edwards.com
TruCCOMS	Catheter in pulmonary artery	High	Mass heat transfer	<1 min	Omega Critical Care, Scotland www.omegacriticalcare.com
OptiQ/Q2Plus	Catheter in pulmonary artery	High	Thermodilution – thermal coil	<5 min	ICU Medical, USA icumed.com
PICCO2	Catheters in central vein and brachial or femoral artery	Medium	Transpulmonary thermodilution + pulse contour analysis	<1 min	Pulsion Medical Systems, Germany www.pulsion.com
LiDCOplus	Catheter in radial or larger artery	Low	Lithium dilution + pulse power analysis using PulseCO algorithm	<1 min	LiDCO, England www.lidco.com
LiDCOrapid	Catheter in radial or larger artery	Low	Pulse power analyis using PulseCO algorithm	<1 min	LiDCO, England www.lidco.com
Vigileo	Catheter in radial or larger artery	Low	Arterial waveform analysis using Flotrac sensor and algorithm	<1 min	Edwards Lifesciences, USA www.edwards.com
FinometerPRO	Probe over finger	None	Modelflow algorithm using finger artery pressure	<1 min	Finapres Medical Systems, Netherlands www.finapres.com
CardioQ-ODM	Probe in oesophagus	Low	Blood flow velocity in descending thoracic aorta by Doppler ultrasound	<1min	Deltex Medical, England www.deltexmedical.com
USCOM 1A	Transducer over thoracic inlet or chest	None	Blood flow velocity at aortic or pulmonary valve by Doppler ultrasound	<1min	Uscom, Australia www.uscom.com.au
NICO2	Sensor at airway or mouth	None	Partial CO_2 rebreathing Fick principle	<5 min	Philips Respironics, USA www.nico.respironics.com
NICOM Reliant	Electrodes on chest wall	None	Bioreactance – phase shifts of propagated electrical signal	<1 min	Cheetah Medical, USA www.cheetah-medical.com

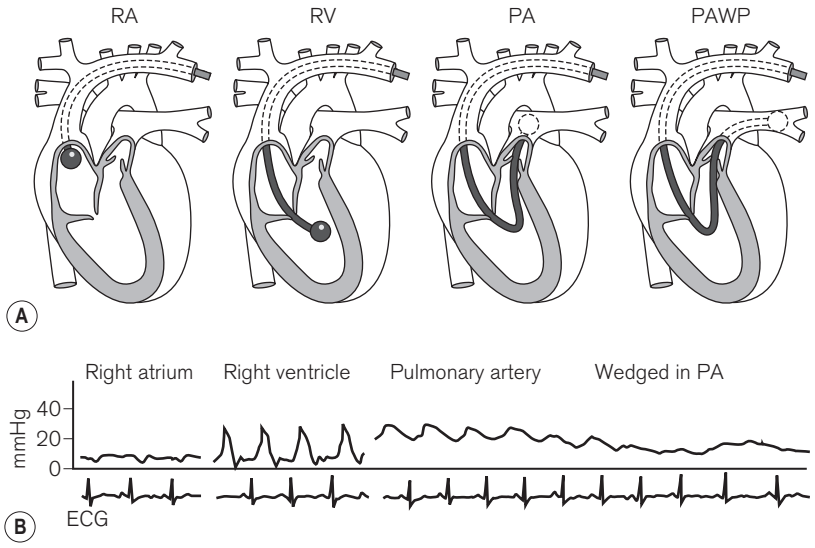


Figure 16.1 **A.** Catheter advancing through right heart chambers and **B.** corresponding pressures in right atrium (RA), right ventricle (RV), pulmonary artery (PA) and wedged (PAWP).

is inflated with air (1.5 ml) and the catheter is advanced with continuous pressure monitoring from the distal lumen. The typical waveforms and pressures encountered as the catheter traverses the right atrium and right ventricle on its way to the pulmonary artery (PA) should be familiar to anyone using this device (Fig. 16.1).

To measure the pulmonary artery wedge pressure (PAWP), which under normal circumstances will approximate left ventricular end diastolic pressures, the catheter must be advanced until it is wedged in one of the medium-sized pulmonary arteries. The balloon is deflated as soon as the PAWP measurement is completed to minimize the risk of causing pulmonary ischemia. A locking mechanism is present at the balloon port to prevent inadvertent inflation of the balloon. Passage of the catheter into the PA depends on an adequate blood flow. If the patient has a very low CO, the catheter may not float into the intended final position, but tend to coil in the right ventricle. In such situations there is a risk that attempts to advance the catheter could literally tie it into a knot.

Cardiac output (CO) with PAC

The PAC measures CO by thermodilution – an adaptation of the indicator-dilution principle according to Stewart and Hamilton. Intermittent CO measurements are made using bolus injections of cold saline or 5% glucose solution as the thermal indicator. Following each injection, a thermodilution curve is obtained by plotting the temperature change of the blood, measured by a rapid-response thermistor at about 4 cm from the distal tip of the catheter. The Stewart-Hamilton equation shows that CO is indirectly proportional to the area under the curve ($\int T_B(t)^{dt}$) and where V = volume of injectate, T_B = blood temperature, T_I = injectate temperature, and K1 and K2 are computational constants. The relationship is given by the formula:

$$CO = \frac{V(T_B - T_I)K_1K_2}{\int T_B(t)^{dt}}$$

Modified pulmonary artery catheters

A PAC equipped with a heating element can measure CO continuously by using a modified thermodilution principle in which the thermal indicator is heat. The catheters made by Edwards Lifesciences and OptiQ (ICU Medical) differ in the way the heating element is constructed and the duration of the heat pulses used to analyze the thermodilution curves. The displayed CO value is the average of intermittent, but frequent measurements performed automatically at intervals of several minutes. *Continuous CO* in this sense is not the same as *beat-to-beat CO*. The averaging time may be prolonged in unstable haemodynamic states or when there is thermal interference following rapid infusion of fluids. Nevertheless, continuous thermodilution CO monitoring is more convenient and may be less prone to operator-dependent errors when compared to intermittent thermodilution.

Data from the continuous CO catheter can be synchronized with the electrocardiogram to obtain right ventricular stroke volume, right ventricular end-diastolic volume, and calculate right ventricular ejection fraction. Oximetric catheters contain a fibre-optic cable and permit

continuous measurement of mixed venous oxygen saturation (SvO_2) from the distal pulmonary artery. Some catheters have embedded pacing wires for atrial, ventricular or dual chamber pacing. Like other central vascular catheters, the PAC can be coated with an antimicrobial agent to reduce the risk of catheter-related infection. A latex-free PAC is also available.

Limitations

1. Central venous cannulation is an invasive procedure with risks of arterial puncture, haemorrhage and pneumothorax.
2. The PAC can become knotted during insertion or later and may need to be removed surgically.
3. The indwelling introducer sheath and PAC are potential causes of sepsis and thrombus formation. The risk of infection appears to increase significantly if the catheters are left in for more than 72 h.
4. The PAC may cause arrhythmias particularly in the presence of hypothermia or electrolyte disturbances.
5. Improper balloon inflation and monitoring (e.g. failure to recognize an inadvertently wedged catheter) can result in arterial rupture, pulmonary infarction or haemorrhage.
6. The presence of the PAC may worsen pre-existing tricuspid regurgitation (TR). Over and underestimation of the true CO have been reported when there is TR. Continuous CO devices may be less affected by TR than when CO is measured intermittently.
7. CO measurement may be inaccurate in the presence of intracardiac shunts.

OESOPHAGEAL DOPPLER METHOD FOR MEASUREMENT OF CARDIAC OUTPUT

Doppler effect

Oesophageal Doppler cardiac output monitors measure blood flow velocity in the descending aorta. An ultrasound probe (see below) is inserted into the distal third of the oesophagus. The ultrasound signal is backscattered by red cells travelling in the descending aorta by virtue of their differing acoustic impedence to the surrounding plasma (see Chapter 31). The returning ultrasound, is at a lower frequency (if flow direction is away from the source) and the difference between the two frequencies – the Doppler shift – is proportional to the velocity of the aortic blood flow as shown in the equation:

$$V = \frac{CF}{2FeCos\phi}$$

where:
V is velocity of red blood cells (blood flow)
C is speed of ultrasound travelling through biological tissues
F is Doppler frequency shift
Fe is emitted frequency from ultrasound device
Cosϕ is cosine of angle (ϕ) between the sound beam axis and the direction of blood flow (*angle of insonation*).

In the following sections the explanation of how this velocity measurement is translated into a calculation of cardiac output is primarily based on the CardioQ-ODM device (Deltex Medical, UK) (see below), which is currently the most widely available oesophageal Doppler cardiac output monitor in the UK. For comparison another device is briefly mentioned at the end.

CardioQ-ODM (Deltex Medical)

In the 1990s Abbott Medical marketed an early version of the oesophageal Doppler monitor, the ODMII, from this manufacturer. A major revision of the monitor resulted in the appearance of the CardioQ in 2000. The current model, the CardioQ-ODM (Fig. 16.2), was introduced in 2009. Compared to the CardioQ, the monitor screen has a higher resolution and a USB has replaced the RS232 ports to facilitate downloading of data and screenshots. The original algorithm to convert flow velocity to stroke volume is still used in the latest model, although some software enhancements have improved the monitor's facility to store and display a variety of haemodynamic data.

Aortic flow signal

Flow signals acquired by the probe are processed by fast Fourier transformation (FFT) and represented on the monitor as a spectral density display of the distribution of

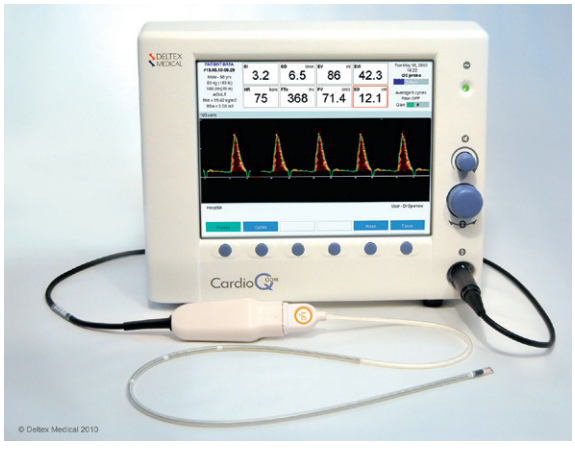


Figure 16.2 The Cardio Q-ODM by Deltex Medical, UK. *Photograph courtesy of Deltex Medical.*

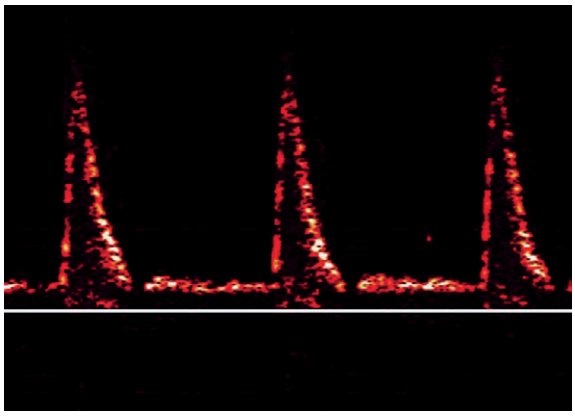


Figure 16.3 Ideal aortic waveform. The brightness of the signal is on the periphery of the waveform.

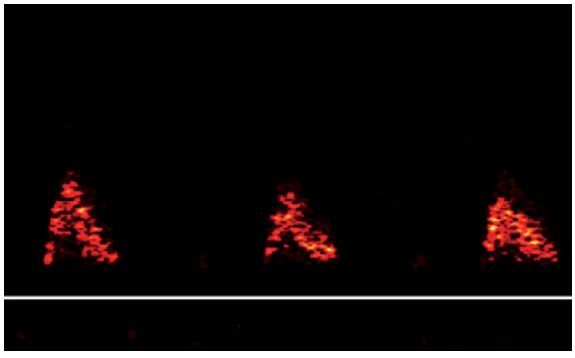


Figure 16.4 Unsatisfactory aortic waveform. This waveform shows spectral dispersion (see text).

red blood cell flow velocities (vertical axis) against time (horizontal axis). Descending aortic blood flow gives a positive deflection and is approximately triangular in shape (Fig. 16.3).

Note that flow in the descending aorta occurs predominantly during systole; there is minimal forward flow in diastole. Correct identification of the descending aortic waveform is a prerequisite for oesophageal Doppler monitoring. Ideally, the aortic waveform should appear triangular with the 'brightness' confined to the peripheral edge and an absence of signal in the central part of the triangle. This type of waveform is characteristic of a plug flow profile (present in the descending aorta) with a narrow spread of red cell velocities. Turbulent flow or unsatisfactory orientation of the transducer towards the descending aorta would show evidence of spectral dispersion (Fig. 16.4).

Signals from other vessels (e.g. pulmonary artery, celiac artery, azygous vein) may be encountered, although these are significantly different in appearance from the descending aortic waveform.

The Doppler frequency is proportional to flow velocity and being in the audible range may be transmitted through speakers. The two rotary controls on the monitor (Fig. 16.2) adjust the volume of the audio output and amplify the signal (gain) seen on the waveform display until the optimal signal is obtained. The same rotary controls may also be used to navigate through various menu options.

The nomogram

The device traces the maximum velocity of the spectrum described above and by calculating the area under the curve a value is derived for *stroke distance*; this being the distance a nominal column of blood moves in the aorta during systole.

The patient's age, weight and height are entered in the monitor and stored in the memory chip of each new probe. The biometric variables are used in a proprietory nomogram to calculate *stroke volume* from the measured stroke distance. The conversion is based on a database of patients who underwent CO monitoring simultaneously with PAC thermodilution and the oesophageal Doppler. From this, the relationship between the velocity time integral (area under the curve) of the aortic flow waveform and stroke volume was formulated. Theoretically, the two are related by the cross-sectional area of the aorta, which is mostly determined by age and size of the individual. In fact the nomogram is in effect a patient-specific calibration constant which also allows for the fact that a significant proportion of cardiac output (brain, coronary and upper limb flow) does not reach the descending aorta. The calibration formula appears to be robust in that it has not had to be modified since its inception over 20 years ago. The accuracy of oesophageal Doppler CO is similar to other less invasive CO devices and considered to be clinically acceptable.

Doppler probe

The CardioQ-ODM uses continuous wave Doppler ultrasound. This is transmitted through an oesophageal probe that consists of a 50 cm length of a tightly coiled, steel spring, mounted with send-and-receive piezo-electric transducers (Fig. 16.5).

The 45° bevel of the transducer defines the angle of insonation relative to the direction of blood flow in the descending aorta. At its proximal end, the probe terminates in an asymmetric connector containing a memory chip. The whole probe is encased in silicone and measures about 19 F – the size of a large nasogastric tube (Fig. 16.6). The adult probe has three markers indicating 35, 40 and 45 cm from the distal end which are used to guide the initial depth of probe insertion via the mouth (35–40 cm) or nose (40–45 cm).

There are several versions of the probe. All are designed for single-patient use only. Adult probes differ in the

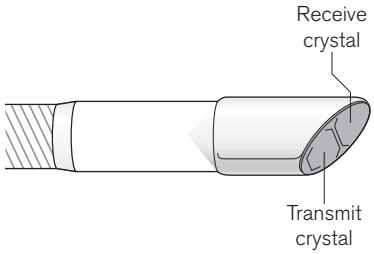


Figure 16.5 The tip of the Cardio Q-ODM probe showing the arrangement of the transducers.



Figure 16.6 Doppler Probe (CardioQ). Note the flexible probe shaft, three depth markers and patient interface connector.

pre-programmed maximum monitoring duration (6, 12 or 240 h). The 'awake Doppler probe' was designed to be softer and more flexible to facilitate insertion through the nose in the awake patient. The paediatric probe has a similar construction, but is shorter and has markings from 15 to 40 cm at 5 cm intervals. This probe has a monitoring limit of 72 h and can be used in children weighing upwards of 3 kg.

Probe insertion

The lubricated probe is inserted into the oesophagus to an appropriate depth. If the probe is inserted orally in an adult, the descending aortic signal is typically found at a depth of 35–40 cm from the lips. To commence monitoring, the probe is connected to the monitor (via a separate patient interface cable) and the patient's age, weight and height are entered. The signal is found by a combination of rotation and slight advancement or withdrawal of the probe. Some experience is required, such that if the user is familiar with the characteristic appearance and sound associated with the optimal aortic flow waveform (Fig. 16.7), then probe positioning and monitoring are rapidly achieved.

Limitations

Readjustment of probe position to ensure optimal aortic waveform is usually required before each measurement. The presence of other devices in the oesophagus (e.g. nasogastric tube) may interfere with signal acquisition. Cardiac output monitoring is unreliable or impossible in some conditions: aortic dissection (turbulent flow and interference due to the intimal flap), coarctation, during cross clamping of descending aorta, and the presence of an intra-aortic balloon. Pharyngo-eosophageal pathology is a relative contraindication for the use of oesophageal Doppler monitoring.

The CardioQ-ODM parameters

In addition to cardiac output (CO) and stroke volume (SV), the CardioQ can display these values indexed to body surface area (cardiac index and stroke volume index). Other key parameters are detailed below.

Corrected flow time (FTc)

Flow time (FT) is the duration of blood flow in the descending aorta resulting from each cardiac ejection. It is indicated by two white triangles at the base of each waveform (Fig. 16.7).

Flow occurs in systole, which occupies approximately one-third of the cardiac cycle. Corrected flow time is inversely related to heart rate and also affected by left ventricular (LV) preload and after-load. Corrected flow time (FTc) indexes the measured FT to a heart rate of 60 bpm and thereby permits interpretation of the influence of loading conditions on FT independent of heart rates. At a heart rate of 60 bpm, the cycle time is 1 s and the corresponding FT is, therefore, approximately one-third of 1 s. In similar manner to adjusting the time interval between ECG waves according to the prevailing heart rate (e.g. 'corrected QT' instead of the QT interval), Bazett's formula is used to calculate FTc from FT. Thus, FTc is obtained from FT divided by the square root of the cycle time (RR interval). The normal FTc is taken as 330–360 ms.

$$FTc = \frac{FT}{\sqrt{Cycle\ time}}$$

A low FTc (<330 ms) is seen in hypovolaemia, decreased LV preload states due to major circulatory 'obstruction' (e.g. pulmonary embolus or isolated right ventricular failure). A low FTc may also reflect an elevated LV afterload, as may occur with significant peripheral vasoconstriction (Fig. 16.8).

A prolonged FTc (>360 ms) is seen with decreased LV afterload, such as in a vasodilated circulation (e.g. sepsis, epidural anaesthesia) (Fig. 16.9).

Peak velocity (PV)

The PV is the highest velocity of descending aortic flow, indicated by the white arrow at the top of the waveform

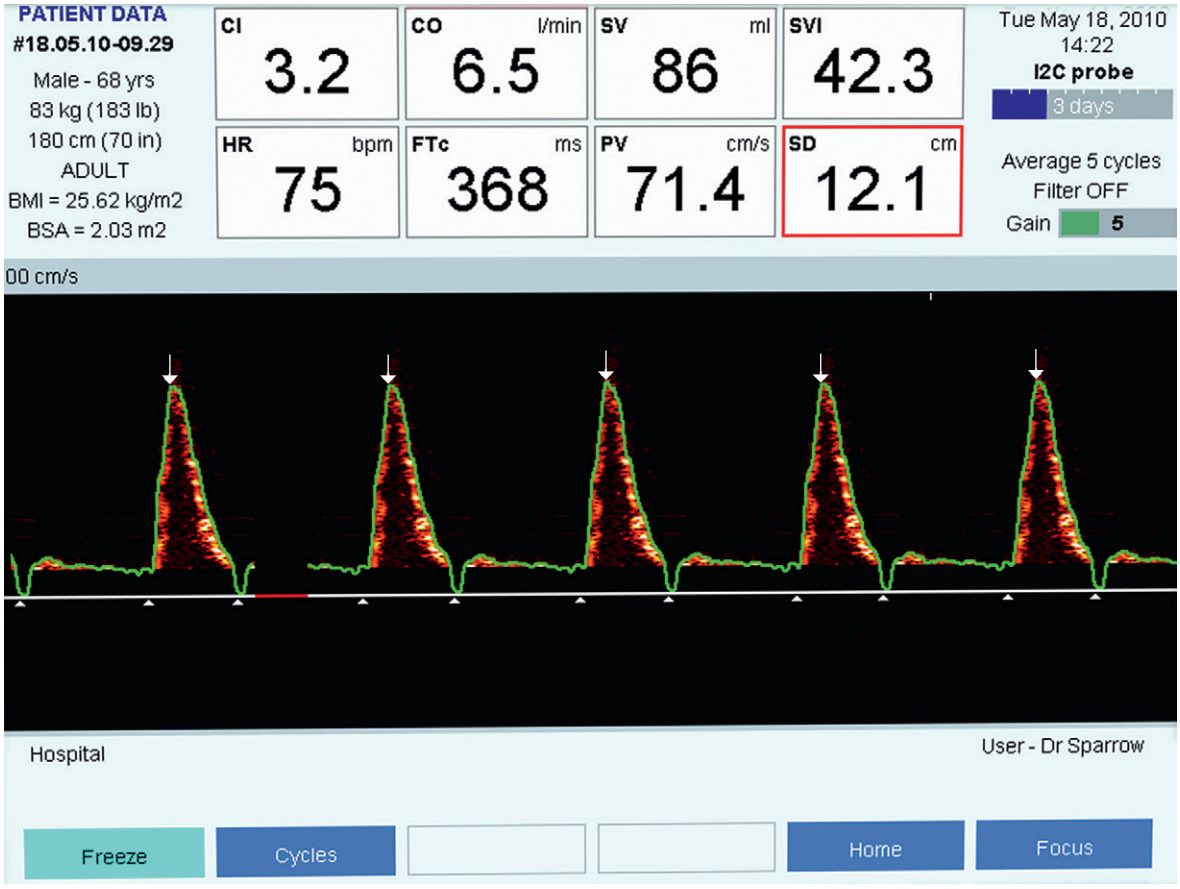


Figure 16.7 Characteristic waveform from a normovolaemic patient. Note the white triangles along the baseline denoting the flow time and white down arrows denoting peak velocity (see text below).

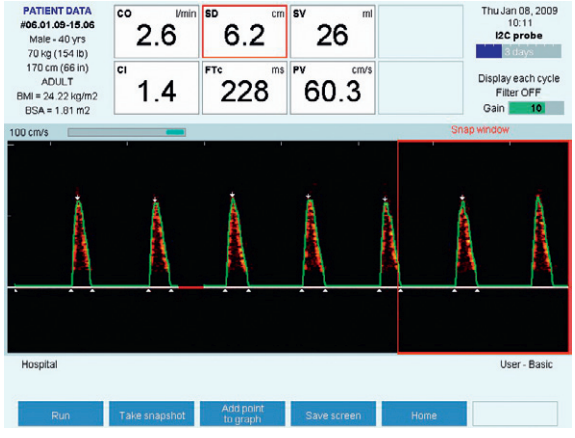


Figure 16.8 Characteristic waveform from a hypovolaemic patient: shortened flow time and low stroke volume.

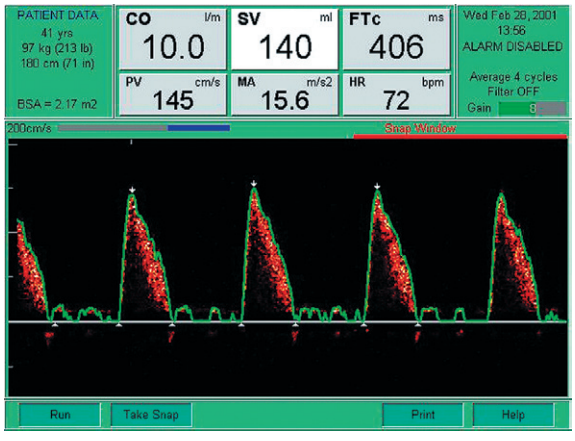


Figure 16.9 Typical waveform from a vasodilated patient: increased stroke volume and flow time longer than 360 ms.

Table 16.2 Peak velocity decreases with age

AGE (YEARS)	PEAK VELOCITY (CM/S)
20	90–120
50	70–100
70	50–80

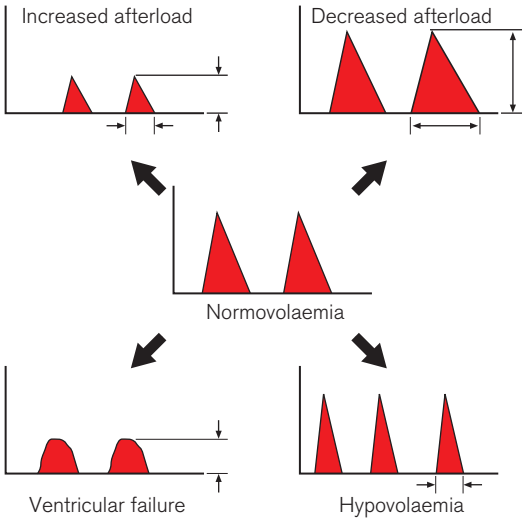


Figure 16.10 Predominant changes observed in the aortic waveform due to alterations in preload, afterload and contractility.

(see Figs 16.7, 16.8 and 16.9). Peak velocity decreases linearly with age, as it is affected by LV contractility, which also declines with age. PV is affected by LV preload (low preload = low PV) and afterload (high afterload = low PV). The normal ranges for peak velocity are shown in Table 16.2. An erroneous, low peak velocity may be recorded if the angle of insonation is significantly greater than 45°. This may occur in the presence of an abnormal anatomical relationship between the axes of the oesophagus and thoracic aorta.

Mean acceleration (MA)

The MA is the average rate of change of velocity of blood flow from the start until the time at peak velocity. Mean acceleration reflects LV contractility, but is also affected by changes in afterload and, to a lesser extent, preload.[6]

Changes in contractility and loading tend to produce characteristic changes in peak velocity or flow time. These are indicated in Fig. 16.10.

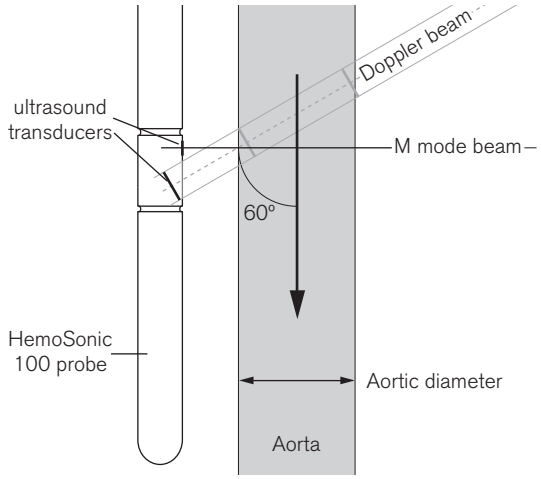


Figure 16.11 Diagram of HemoSonic 100 probe showing the angled and perpendicular ultrasound beams.

HemoSonic 100, Arrow Critical Care Products, USA

The HemoSonic system utilizes a probe with two ultrasound transducers (Fig. 16.11). One points perpendicularly to the axis of the probe and is used in M Mode (see Chapter 31) to confirm probe position and to measure aortic diameter. The second distally placed pulsed Doppler transducer is angled at 60° towards descending aortic flow to measure flow velocity in the same area. The probe is a little larger as a consequence and is reused, necessitating a disposable cover and oral usage only. This device is no longer in production but illustrates an alternative approach.

ARTERIAL PRESSURE WAVEFORM ANALYSIS

Devices that estimate CO from the blood pressure should be very appealing. As fluctuations of arterial blood pressure around a mean value that occur with each ventricular contraction are caused by the *volume* of blood that is ejected into the arterial system, an inference can be made as to the cardiac output. However, it does require beat by beat direct measurement of arterial pressure. As direct arterial pressure monitoring is only minimally invasive, is performed routinely and is relatively devoid of complications, this has now become a feasible option.

The increase in aortic pressure that occurs with left ventricular ejection will depend on the stroke volume, aortic capacitance and the vascular tone or compliance. The CO estimated from arterial pressure analysis can be calibrated directly against a measured reference CO or

inferred using reference data biometrically matched to the patient. The limitations of these devices reflect the fundamental problem of inferring volume from pressure. Pressure-derived CO tends to become inaccurate when there are rapid or extreme changes in peripheral resistance. Cardiac output estimation may also be unreliable in the presence of rapid or persistent arrhythmia, if the arterial cannula becomes kinked, if the waveform is damped, and when an intra-aortic balloon is being used.

Manufacturers use a variety of usually proprietary algorithms for deriving cardiac output from analysis of the arterial waveform; they are discussed below briefly under each device.

PiCCO2 (Pulsion Medical Systems)

The first arterial pressure-derived CO monitor (PiCCO) from Pulsion Medical Systems was available in 1997. The PiCCO2 was brought out in 2008 and uses a similar algorithm to the original monitor. Stroke volume is estimated from the area under the systolic portion of the arterial pressure waveform with adjustments for aortic compliance, impedance and the shape of the arterial waveform (Fig. 16.12).

The treatment of the arterial waveform in this way is best described as '*pulse contour analysis*'. To calculate the systolic area, the waveform must be from a central artery from where the dicrotic notch can be detected more reliably. A thermistor-tipped cannula is used to monitor pressure and temperature in the femoral artery. The brachial and axillary arteries are acceptable alternatives. Depending on the artery selected and the size and age of the patient, the recommended cannula can range between 3 and 5 F in size and 7 and 22 cm in length. The radial artery is not generally suitable, although a long (35 cm) catheter is available as an option – its tip would lie in the brachial artery.

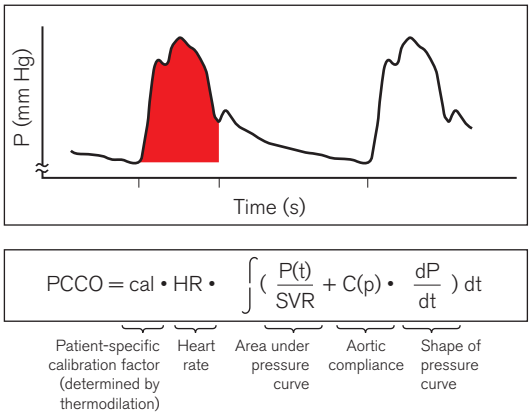


Figure 16.12 Cardiac output calculation in the PiCCO. PCCO, pulse contour cardiac output; SVR, systemic vascular resistance.

Transpulmonary thermodilution

Pulse contour data (above) is calibrated using CO measured by intermittent transpulmonary thermodilution, which is based on the same principle as PA thermodilution. A bolus (15 ml in most adults) of cold (<8 °C) saline is injected rapidly (within 7 s) into a central venous catheter and the temperature change recorded at the arterial cannula is plotted. Compared to PA thermodilution, most studies have concluded that transpulmonary thermodilution yields comparable CO values. The temperature change must be measured in a central artery and this is another reason for cannulating the femoral in preference to more peripheral arteries. The average of three CO values is used to calibrate the pressure data. The CO displayed on the monitor is the average of CO values in the previous 12 s. It is accurate as long as recalibration is performed at least every 8 h.

Additional variables

One of the unique features of the PiCCO2 is its ability to estimate cardiac preload, cardiac function and extravascular lung water. After injection into a central vein, the cold saline indicator is distributed serially through three compartments – (a) in the thorax, (b) the right heart, (c) the lungs and the left heart – before reaching the artery where the temperature change is measured. The volumes of these compartments can be calculated with reference to the transpulmonary thermodilution curve. The mean transit time (MTt) is the time taken for half of the indicator to be detected (Fig. 16.13).

The product of MTt and CO equals the intrathoracic thermal volume (ITTV). The slope of the terminal portion of the thermodilution curve is governed by the largest of the three compartments – the lungs in this case.[7] The product of the exponential downslope time (DSt) and CO

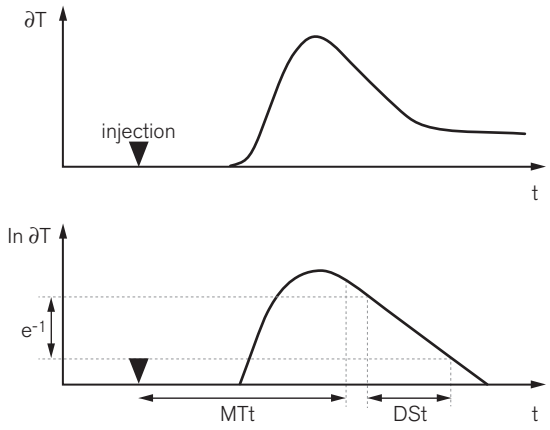


Figure 16.13 Transpulmonary thermodilution curve for PiCCO2.

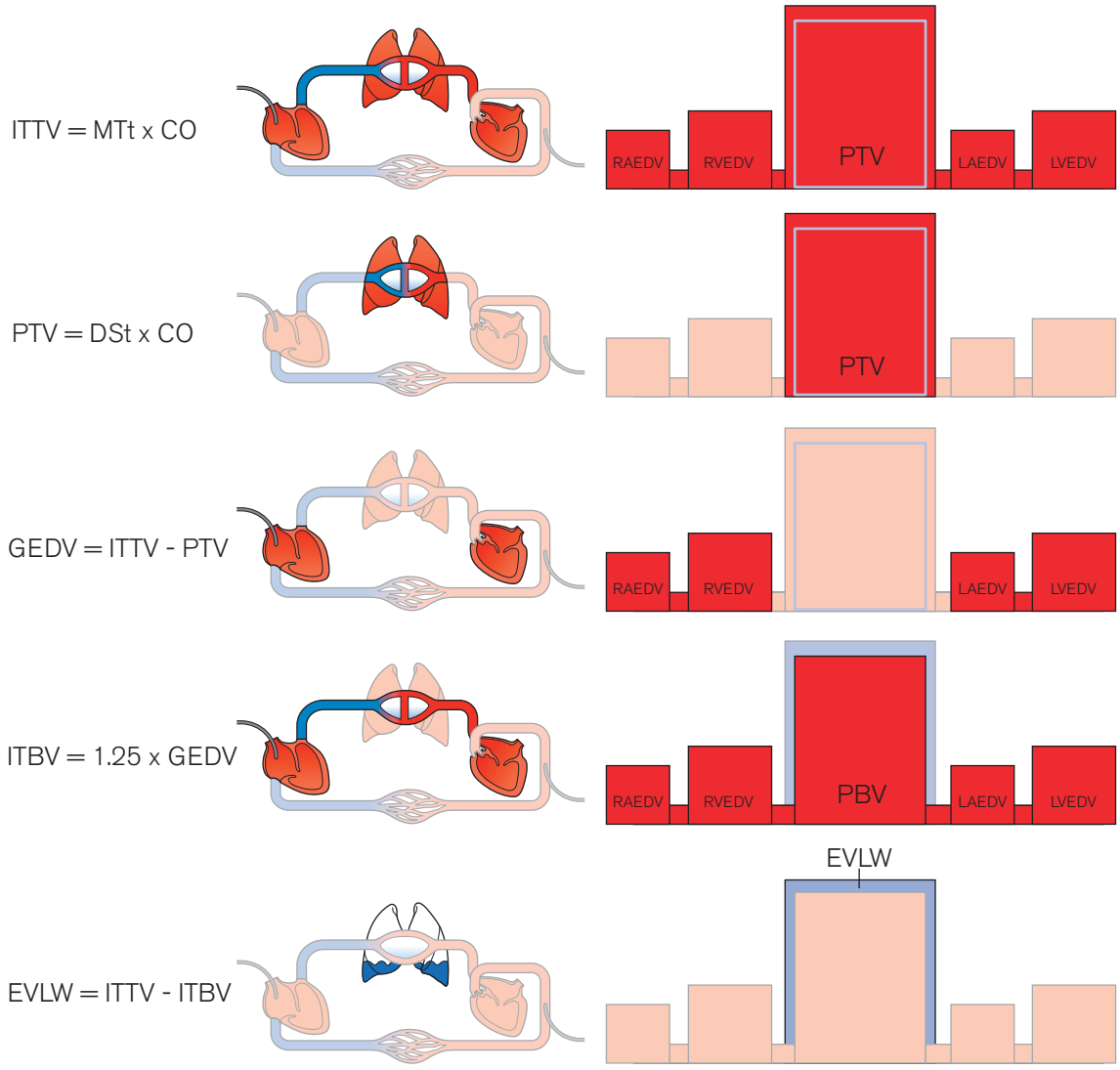


Figure 16.14 Schematic representation of the calculation of various volumes in transpulmonary thermodilution. RAEDV, LAEDV: right and left atrial end diastolic volumes. RVEDV, LVEDV: right and left ventricular end diastolic volumes. PTV: pulmonary thermal volume. PBV: pulmonary blood volume. EVLW: extravascular lung water. GEDV: global end diastolic volume. ITTV: intrathoracic thermal volume. ITBV: intrathoracic blood volume. MTt: mean transit time, DSt: down stroke time. CO: cardiac output.

gives the pulmonary thermal volume (PTV). The volume of blood in the right and left heart chambers at the end of diastole, called the global end diastolic volume (GEDV), is the difference between PTV and ITTV. The GEDV gives a relative measure of cardiac preload and at least one study suggests that absolute values may depend on gender and age.[8] Based on previous work showing that the intrathoracic blood volume (ITBV) is consistently 25% greater than the GEDV, extravascular lung water (EVLW) is the difference between ITTV and ITBV.[9] These volume calculations are summarized in Fig. 16.14. Other haemodynamic indices can be calculated from the available data and include the global ejection fraction (GEF), cardiac function index (CFI), cardiac power (CP) and pulmonary vascular permeability index (PVPI).

LiDCO*plus* and LiDCO*rapid*

These monitors use the same algorithm (PulseCO) to estimate CO from the arterial pressure waveform. The

arterial pressure waveform is slaved to the LiDCO devices via standard analogue outputs from any patient monitor. In the LiDCO*plus*, CO is also measured intermittently by lithium-dilution (see below) and this value is used to calibrate the nominal CO obtained by the algorithm.

Lithium dilution cardiac ouput

The principle of measuring CO by lithium-dilution is the same as for other indicator-dilution methods. A bolus of lithium chloride is injected into a peripheral or central vein. Lithium concentration is measured in blood from a peripheral artery by a sensor containing a lithium-selective electrode. The sensor is prepared by flushing its flow-through cell with saline before connecting it, via a three-way tap, to any existing arterial line (Fig. 16.15). The saline soaks the internal wick that makes the electrical connection between the blood in the cell and a reference electrode. To measure CO, blood is drawn from the arterial line at a constant rate (4.5 ml min^{-1}) by a battery-operated pump. Once the baseline voltage of the sensor stabilizes, a 2 ml bolus of lithium chloride (0.15 mM/ml) is rapidly flushed into a vein. The voltage response of the sensor is related to the change in lithium concentration; hence, a plot of lithium concentration against time can be obtained and used to calculate CO (Fig. 16.16). Because the lithium is only distributed in the plasma, the haemoglobin concentration is necessary to convert plasma lithium concentration to whole blood concentration in order to derive blood flow.

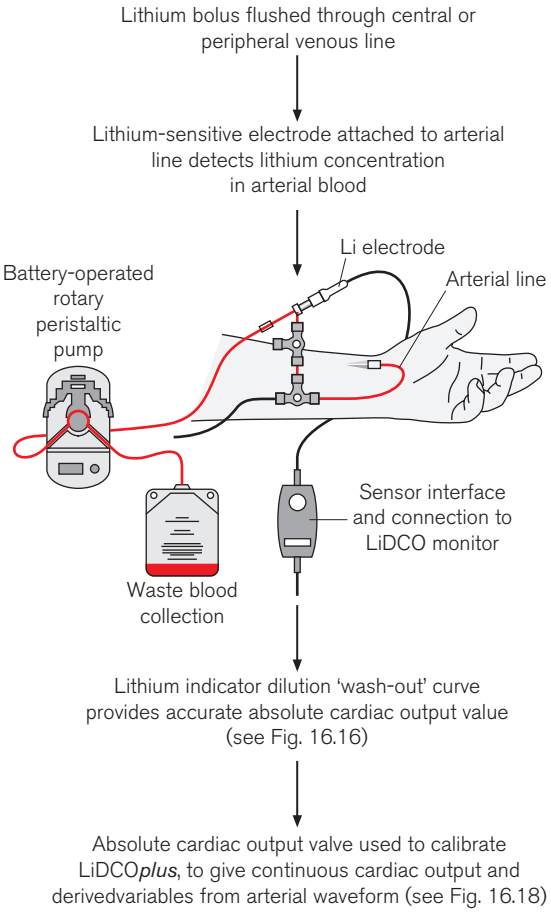


Figure 16.15 The set up for the LiDCO*plus* Haemodynamic Monitor.

As in other indicator dilution methods, intracardiac shunts and valvular regurgitation can lead to erroneous CO values. This method of measuring CO cannot be used in patients on lithium therapy in whom baseline concentrations are significantly above those seen when lithium is used as an indicator. Significantly in the anaesthesia setting: the metabolites of the non-depolarizing muscle relaxants affect the lithium electrode and hence the timing of their injection in relation to performing cardiac output measurement for calibration needs to be considered. Long-acting agents have an effect for 20 min, short-acting agents for up to 2 h.

Because of potential toxicity implications, LiDCO*plus* monitoring is not recommended in patients under 40 kg or during pregnancy.

PulseCO continuous cardiac output

The LiDCO*plus* and LiDCO*rapid* use the PulseCO algorithm to derive CO. In essence, the algorithm attempts to convert a pressure signal into a volume signal in order to determine stroke volume.

The pressure waveform data sampled at 100 Hz is compliance corrected based on cadaveric aortic pressure/volume data and scaled for individual aortic capacitance to produce a corresponding volume waveform in the same time domain. Modified auto covariance (a type of signal processing analysis) is then applied to derive the input function for this waveform, i.e. the stroke volume (Fig. 16.17).

The algorithm is calibrated in the LiDCO*plus* using the CO value obtained by lithium-dilution. The calibration factor is in effect the ratio of the nominal and lithium-dilution CO values. The calibration factor reflects the aortic capacitance, which is patient-specific and assumed to remain relatively unchanged over a period of a few days. However, depending on the underling hemodynamic condition, the frequency of calibration may need to be as frequently as every 4 h.[10]

Because the algorithm is not based on waveform morphology (in contrast to 'pulse contour analysis'), it is less impacted by wave reflection or damping, making a peripheral artery (e.g. radial) generally suitable as a site for pressure monitoring. The treatment of the arterial pressure data in this way has been termed *'pulse power analysis'*.

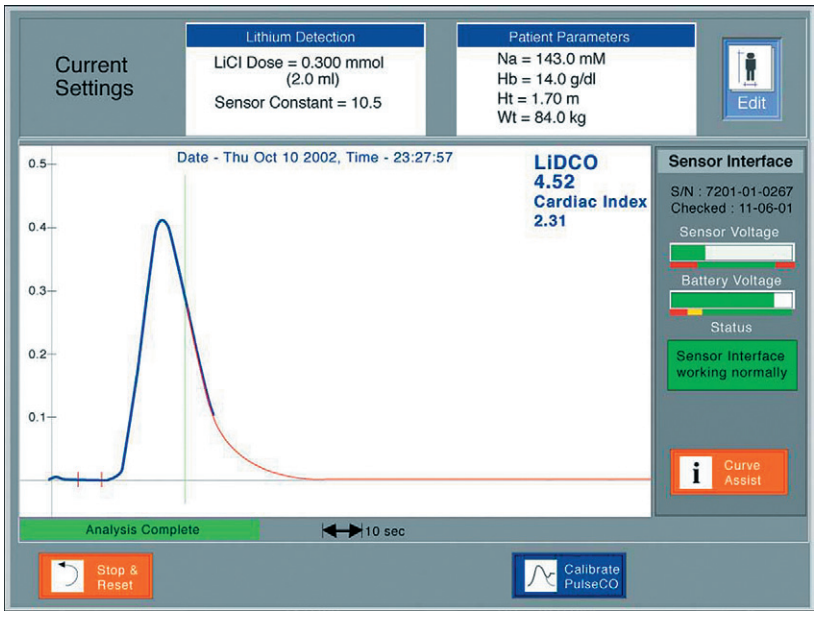
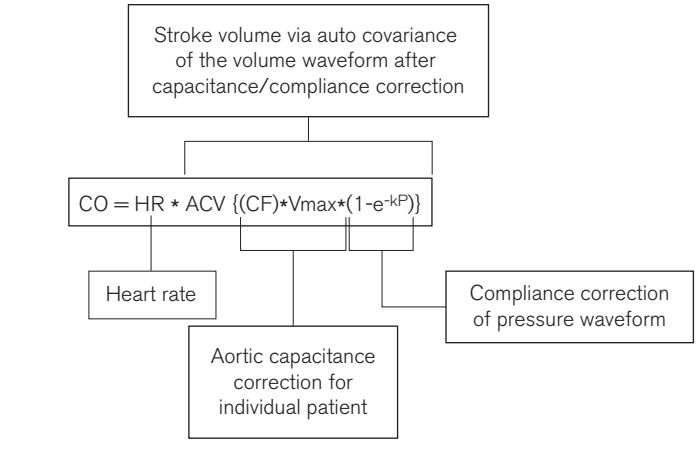


Figure 16.16 LiDCO*plus* Haemodynamic Monitor display of the CO value, which is used to calibrate the waveform analysis.



CO = Cardiac output
HR = Heart rate
ACV = Auto co variance function
CF = Calibration factor
Vmax = 250ml
K = Compliance correction coefficient
P = Arterial pressure, mmHg

Figure 16.17 Cardiac output calculation from 'pulse power analysis' in the LiDCO monitor.

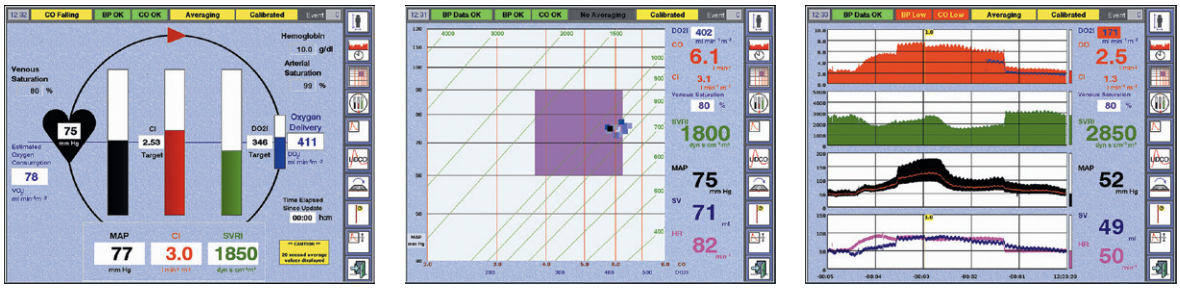


Figure 16.18 Three screen shots of different configurations of the LiDCO*plus* display.

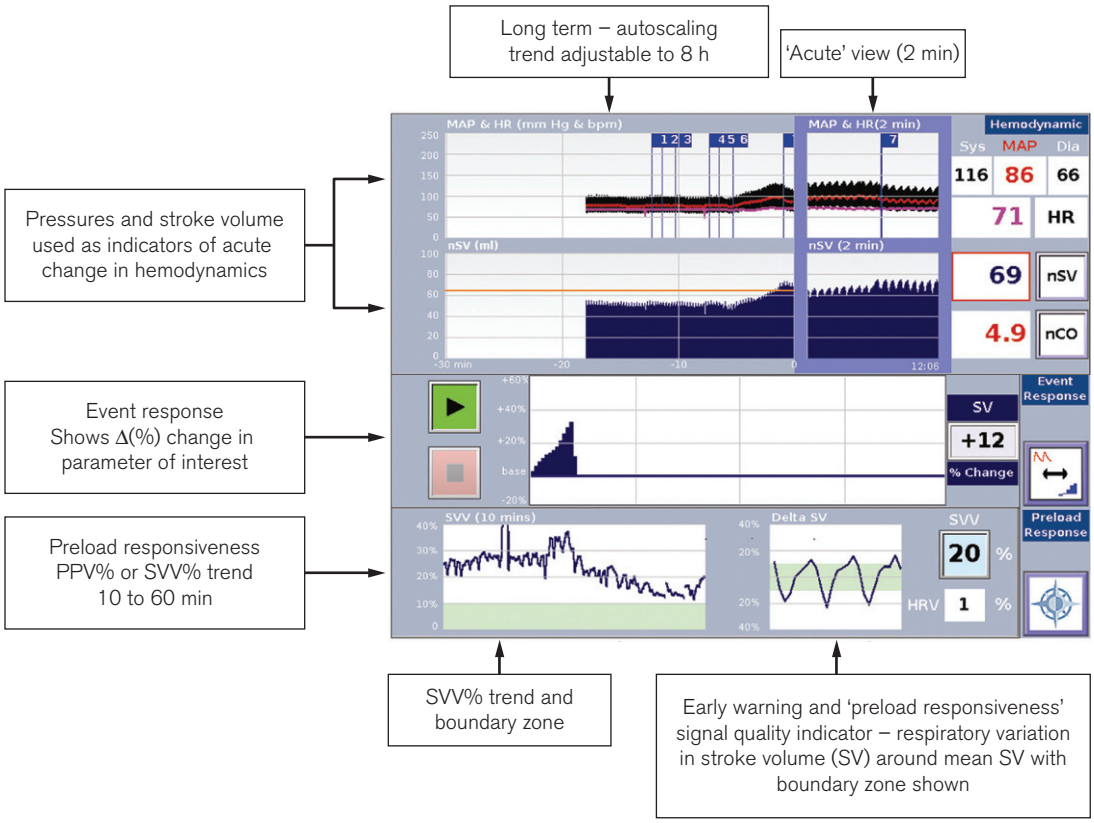


Figure 16.19 Annotated screen shot of the LiDCO*rapid*.
Courtesy of LiDCO Ltd, UK.

The LiDCO*plus* screen can be configured to display the haemodynamic data as trends or graphically to assist decision-making (Fig. 16.18).

LiDCO*rapid*

This device is intended to be minimally invasive and has been designed specifically for intraoperative monitoring. It has a single comprehensive screen displaying a combination of preload indicators such as pulse or stroke volume variation and standard haemodynamic trends (Fig. 16.19). A credit-card style smart card slots into the side of the monitor to activate it for use. The monitor requires only the arterial pressure to be slaved from the patient monitor. The PulseCO algorithm is used to convert arterial pressure to a nominal CO. Using *in vivo* calibration data previously obtained in postoperative patients, the nominal CO is converted to an estimated CO adjusted for the patient's age, weight and height. The resulting CO values are known to be less accurate than those from



Figure 16.20 The FloTrac sensor.
Photograph courtesy of Edwards Lifesciences Corporation, Irvine, California.

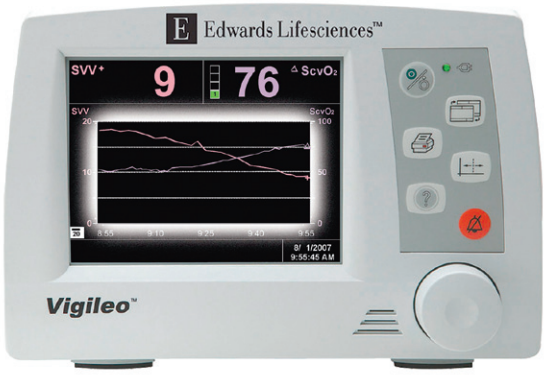


Figure 16.21 The Vigileo monitor.
Photograph courtesy of Edwards Lifesciences Corporation, Irvine, California.

LiDCO*plus* calibrated using lithium-dilution, but trends in estimated variables and observed responses to therapeutic interventions should hold as true as for calibrated CO. The LiDCO*rapid* can also be calibrated using a reference CO obtained from another source.

FloTrac-Vigileo (Edwards Lifesciences, USA)

The FloTrac sensor in conjunction with the Vigileo monitor continuously estimates CO from a peripherally measured arterial pressure. The standard arterial pressure transducer is replaced by the FloTrac sensor (Fig. 16.20), which is connected to a cannula in the femoral (preferred), brachial or radial artery. The sensor and associated pressure transducer tubing are packaged for single-patient use. After zeroing the sensor to atmosphere the patient's age, gender, height and weight are entered in the Vigileo (Fig. 16.20). The patient bedside monitor and the Vigileo are both connected to the FloTrac so that the arterial pressure can be monitored in the usual fashion while the data are used simultaneously to derive CO. As for any arterial pressure-derived CO monitor, under- or over-damping of the arterial waveform should be avoided by ensuring that the cannula is not malpositioned and the lumen remains patent.

FloTrac algorithm

The FloTrac algorithm for '*arterial pressure based cardiac output*' is given as:

$$CO = PR \cdot \delta AP \cdot \chi$$

where PR is the transduced pulse rate and δAP. χ represents stroke volume.

Stroke volume is derived from a complex process of statistical analysis and signal processing.[11] In essence, stroke volume is determined by two elements: the amplitude of arterial pressure and the characteristics of the arterial tree. The first element is represented by the standard deviation of arterial pressure (δAP), which is calculated from a large number of arterial pressures sampled at 100 Hz over 20 s (i.e. about 2000 values). The second element, denoted by 'chi' (χ), is a dynamic polynomial function accounting for arterial compliance and vascular tone. Arterial compliance is estimated from the patient's age, mean arterial pressure, gender and size. The skewness and kurtosis of the arterial waveform is also analyzed to determine changes in vascular tone. Since it was first introduced in 2005, the FloTrac algorithm has undergone two major revisions. In the second-generation software, the χ function was updated more frequently at 1-min instead of 10-min intervals. The current third-generation software, launched in 2009, improved the accuracy of CO estimation in patients with low systemic vascular resistance (e.g. sepsis).

The Vigileo monitor (Fig. 16.21) has the capability to measure central venous oxygen saturation ($ScVO_2$) continuously using the PreSep central venous oximetry catheter (Fig. 16.22). $ScVO_2$ has been shown to correlate reasonably well with mixed venous oxygen saturation (SvO_2) and, therefore, can be used to assess the adequacy of oxygen delivery. The Vigileo displays stroke volume variation as a predictor of preload responsiveness during positive pressure ventilation. The current FloTrac algorithm cannot be used in adults below 40 kg.

The NICOM

The NICOM (Cheetah Medical Inc, USA) is a completely *n*on-*i*nvasive *CO m*onitor. Four dual electrode pads are

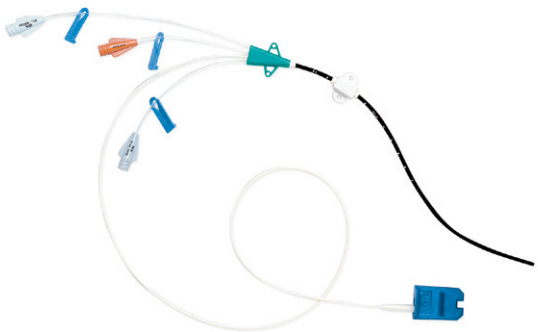


Figure 16.22 PreSep central venous oximetry catheter. *Photograph courtesy of Edwards Lifesciences Corporation, Irvine, California.*

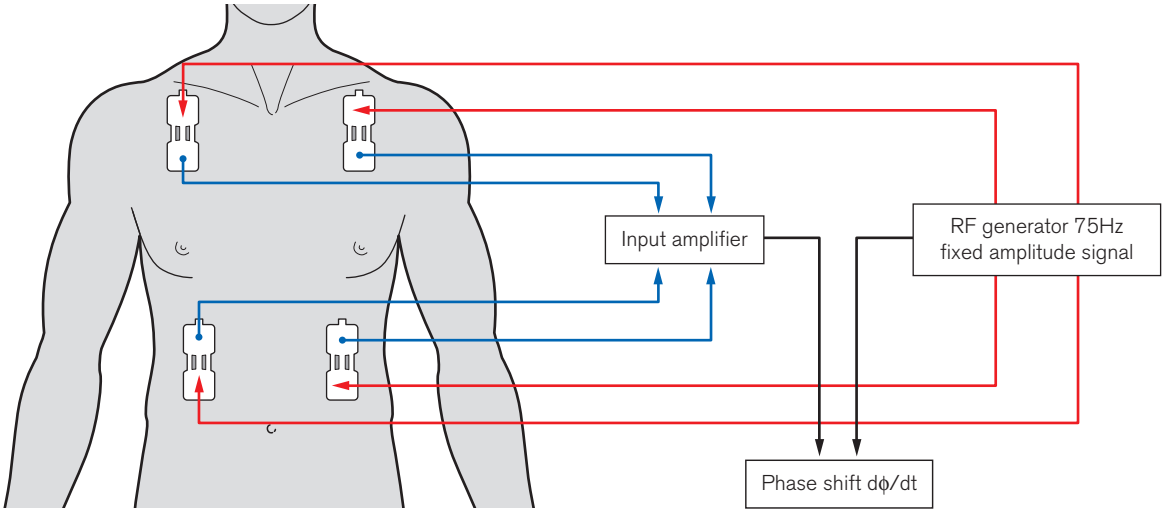


Figure 16.23 Schematic showing the electrode placements and operating principles of the NICOM device.

applied symmetrically on the thorax, approximately at mid-clavicle above and at the lower margin of the rib cage below (Fig. 16.23). Each electrode pad is positioned so that the two sections of the electrode are orientated vertically. The exact location of the electrode is not critical and they can also be placed on the back of the thorax. Cardiac output is measured using the proprietary Bioreactance method. A 75 KHz radiofrequency signal is transmitted across the thorax from the outer section of the electrodes. The propagated signal is sensed by the inner sections of the electrodes located at the opposite side of the thorax. Pulsatile changes in aortic blood volume in the thorax, which are correlated to the stroke volume, cause changes in amplitude of the propagated electrical signal. Changes in intrathoracic volume also produce changes in electrical capacitative and inductive properties (reactance), which are seen as phase shifts between the applied and sensed signal.[12] Traditional bioimpedance measures the change in the amplitude of the propagated signal. Bioreactance measures the change in phase (Fig. 16.24). Because of this, signal-to-noise ratio is improved in the NICOM compared to thoracic bioimpedance-based devices. In practical terms, the NICOM is less affected by the patient's body shape, movement or the exact location of the electrodes on the thorax. Potential interference from other radiofrequency signals in the clinical environment can also be filtered out without loss of the target signal.

There are built-in connectors for oscillotonometric blood pressure measurement. In addition to CO, which is updated at either 0.5 or 1.0 min intervals, the monitor

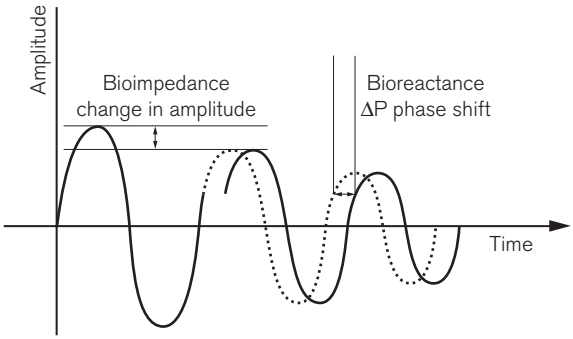


Figure 16.24 Signal analysis for bioimpedance and Bioreactance.
Diagram courtesy of Cheetah Medical Inc.



Figure 16.25 NICOM.
Photograph courtesy of Cheetah Medical Inc.

will also calculate peripheral resistance, cardiac power, stroke volume variation, oxygen delivery and thoracic fluid content (derived from the measured impedance). The parameters to be displayed on the monitor can be customized by the user (Fig. 16.25). These features and non-invasive CO monitoring allow the NICOM to be used in a wider variety of clinical environments than most of the currently available CO devices.[13]

REFERENCES

1. Swan HJ, Ganz W, Forrester J, Marcus H, Diamond G, Chonette D. Catheterization of the heart in man with use of a flow-directed balloon-tipped catheter. *NEJM* 1970;**283**:447–51.
2. Richard C, Warszawski J, Anguel N, Deye N, Combes A, Barnoud D, et al. Early use of the pulmonary artery catheter and outcomes in patients with shock and acute respiratory distress syndrome: a randomized controlled trial. *JAMA* 2003;**290**:2713–20.
3. Harvey S, Harrison DA, Singer M, Ashcroft J, Jones CM, Elbourne D, et al. Assessment of the clinical effectiveness of pulmonary artery catheters in management of patients in intensive care (PAC-Man): a randomised controlled trial. *Lancet* 2005;**366**:472–7.
4. Binanay C, Califf RM, Hasselblad V, O'Connor CM, Shah MR, Sopko G, et al. Evaluation study of congestive heart failure and pulmonary artery catheterization effectiveness: the ESCAPE trial. *JAMA* 2005;**294**: 1625–33.
5. The National Heart Lung and Blood Institute Acute Respiratory Distress Syndrome (ARDS) Clinical Trials Network, Wheeler AP, Bernard GR, Thompson BT, Schoenfeld D, et al. Pulmonary artery versus central venous catheter to guide treatment of acute lung injury. *New Eng J Med* 2006;**354**:2213–24.
6. Kong R. Transoesophageal diagnostic Doppler monitoring. In: Arthurs G, Nicholls B, editors. *Ultrasound in anesthetic practice*. Cambridge: Cambridge University Press; 2008. p. 58–66.
7. Newman EV, Merrell M, Genecin A, Monge C, Milnor WR, McKeever WP. The dye dilution method for describing the central circulation. An analysis of factors shaping the time-concentration curves. *Circulation* 1951;**4**:735–46.
8. Wolf S, Riess A, Landscheidt JF, Lumenta CB, Friederich P, Schürer L. Global end-diastolic volume acquired by transpulmonary thermodilution depends on age and gender in awake and spontaneously breathing patients. *Crit Care* 2009; **13**:R202.
9. Effros RM, Pornsuriyasak P, Porszasz J, Casaburi R. Indicator dilution measurements of extravascular lung water: basic assumptions and observations. *Am J Physiol Lung Cell Mol Physiol* 2008;**294**:L1023–31.
10. Cecconi M, Fawcett J, Grounds RM, Rhodes A. A prospective study to evaluate the accuracy of pulse power analysis to monitor cardiac output in critically ill patients. *BMC Anesthesiol* 2008;**8**:3.
11. Pratt B, Roteliuk L, Hatib F, Frazier J, Wallen RD. Calculating arterial pressure-based cardiac output using a novel measurement and analysis method. *Biomed Instrum Technol* 2007;**41**:403–11.
12. Keren H, Burkhoff D, Squara P. Evaluation of a noninvasive continuous cardiac output monitoring system based on thoracic bioreactance, American Journal of Physiology. *Am J Physiol Heart Circ Physiol* 2007;**293**: H583–9.
13. Squara P, Denjean D, Estagnasie P, Brusset A, Dib JC, Dubois C. Noninvasive cardiac output monitoring (NICOM): a clinical validation. *Intensive Care Med* 2007;**33**:1191–4.

FURTHER READING

Rhodes A, Sunderland R. Arterial pulse power analysis: the LiDCOplus system. In: Pinsky M, Payen D, editors. *Functional haemodynamic monitoring*. Berlin: Springer-Verlag; 2005. p. 183–92.

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Chapter 17

Depth of anaesthesia and neurophysiological monitoring

Andrew Morley and Gustav Strandvik

CHAPTER CONTENTS

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Neurological function may be monitored for many reasons in anaesthesia and intensive care. Nerve stimulation can help with pharmacological neuromuscular blockade, anatomical localization during peripheral nerve blockade and confirmation of the integrity of neural pathways during spinal surgery. Analysis of cortical electrical activity is the underlying principle of most devices which purport to reflect anaesthetic depth. Finally, some monitors are available which may reduce the risk of cerebral ischaemia and neurological deficit in particular surgical circumstances where oxygen delivery to the brain may be compromised.

NERVE STIMULATORS

Nerve stimulators are designed to administer electrical stimuli to peripheral nerves. The responses may be assessed in different ways. The strength of an electrical stimulus applied to a nerve is defined by its *charge* (coulombs, Q). This is equivalent to the product of the *current* passed (amperes, A) and the *duration* (seconds, s) of that pulse of current.

Most nerve stimulators employ constant current circuitry, in which the difference between the current set by the user and the actual current delivered is detected and the applied voltage is automatically adjusted to minimize any disparity. This situation may arise when a dried-out skin electrode causes high impedance and consequently a low initial current flow.

Assessment of neuromuscular blockade

During assessment of neuromuscular blockade a *supramaximal stimulus* (i.e. one in which all axons in the nerve are made to discharge) is administered to a motor nerve through a pair of ordinary adhesive silver/silver chloride electrodes, applied to the overlying skin. Typically, the stimulus pulse duration might be 0.2 ms, the current 60 mA and the resulting charge, 12 μC. With pulses shorter than 0.2 ms, the amplitude required to generate a supramaximal stimulus may exceed the capacity of the stimulator, while pulses longer than 0.3 ms run the risk of repetitive nerve or muscle stimulation. Higher charges than are classically used may be required to guarantee supramaximal stimulation in nerves with prolonged conduction times (e.g. in diabetic neuropathy).[1] The charge required to deliver a supramaximal stimulus is less when the negative electrode is placed distally.[2]

Basic hand-held devices provide a stimulus with a fixed current and pulse width and permit a limited range of stimulation patterns (Fig. 17.1). Other models have liquid crystal screens and user-variable current, enabling both neuromuscular monitoring and nerve localization.

The motor response evoked by peripheral nerve stimulation may be measured in different ways, but is typically evaluated visually. Usually, contraction of adductor pollicis is observed following ulnar nerve stimulation at the wrist. The facial nerve in the cheek is also used, but direct muscle stimulation may contribute to the resulting contraction. More advanced devices attempt to *quantify* the

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Figure 17.1 RS6 neuromuscular monitor (G. Rutter Ltd). Current (60 mA) and pulse width (0.25 ms) are fixed. A train-of-four, double-burst or tetanic stimulus may be selected. The leads may be removed to reveal probes (as shown) for direct application to moistened skin.

(force of) muscular contraction elicited by peripheral nerve stimulation.

Currently available monitors allowing objective assessment of the motor response to neural stimulation use *accelerometric* or *kinemyographic* technology.

Accelerometry/acceleromyography

The basis of acceleromyography is that force is the product of mass and acceleration.[3] During the assessment of neuromuscular blockade, mass is relatively unchanged so acceleration is proportional to force. An accelerometer comprises a small mass suspended on a strain gauge within a 'box' attached to the accelerating object to be studied, the acceleration then being derived from the force exerted on the strain gauge.

A number of accelerometric devices are available, such as the TOF-Watch (Organon Teknika, Fig. 17.2).

Piezoelectric methods ('Kinemyography')

When compressed or distorted, piezoelectric materials produce a charge proportional to the degree of alteration in shape. The M-NMT module for the S/5 anaesthesia monitor (GE Healthcare) uses a piezoelectric sensor incorporated into a clip placed on the patient's thumb and index finger (Fig. 17.3). Thumb movement on stimulation is converted to an electrical signal and displayed graphically as a proportion of the maximal response. Monitoring, therefore, needs to be started before neuromuscular blockade but after anaesthetic induction. The module initially determines the current required for a supramaximal stimulus and sets a reference response level. By using different patient interface attachments the module can also be used for electromyographic recording or for nerve localization.

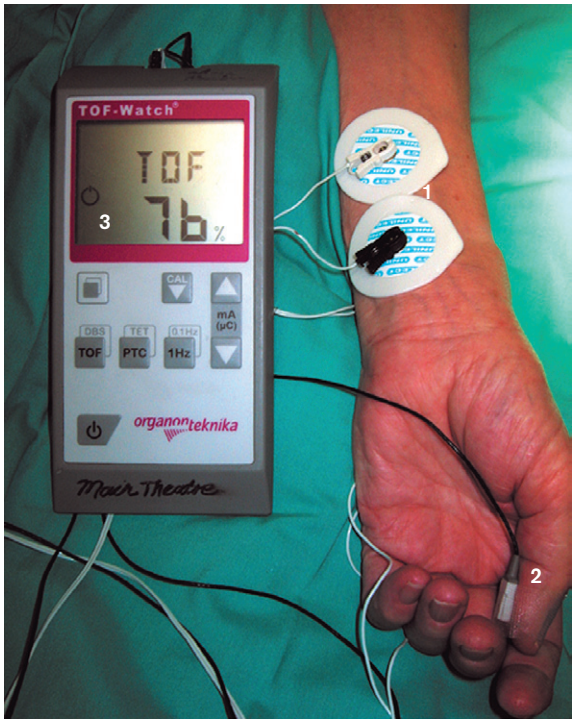


Figure 17.2 The TOF-Watch monitor (Organon Teknika). Stimulating current from 0–60 mA is set by the user. Accelerometric assessment of neuromuscular blockade or nerve localization may be performed. For the former, the device is calibrated after induction and before administration of muscle relaxant. (1) Stimulating electrodes applied over ulnar nerve; (2) accelerometric transducer taped to thumb; (3) display screen indicating train-of-four ratio: when less than four responses are detected, the number is indicated instead.

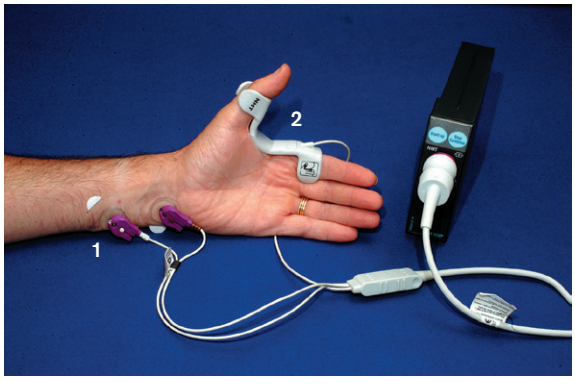


Figure 17.3 The mechanosensor for the Datex–Ohmeda M-NMT neuromuscular transmission module. (1) Stimulating electrodes applied over ulnar nerve; (2) piezoelectric mechanosensor.

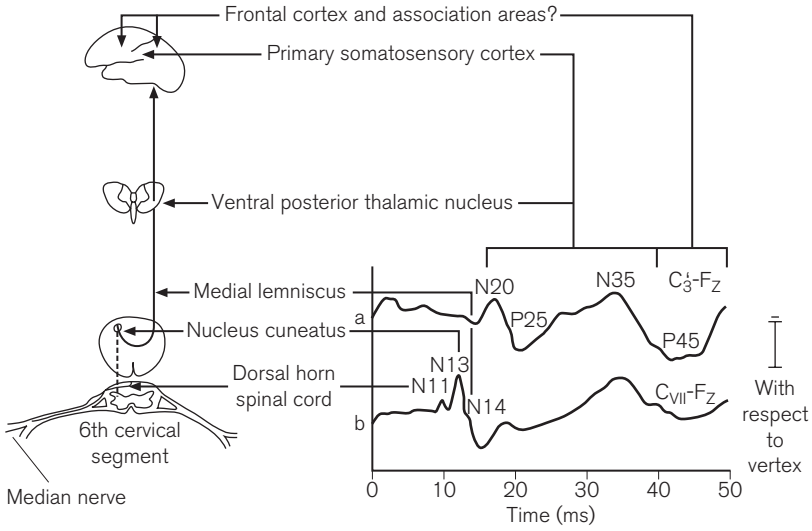


Figure 17.4 Somatosensory evoked response. The stimulating electrode is over the median nerve. Recording electrodes are over **A.** the somatosensory cortex and **B.** the seventh vertebra. The sequential peaks and troughs in the evoked response are named according to their latency (ms) and direction. Particular periods of the response correspond to specific neuroanatomical regions as shown.

(Thornton C, Sharpe RM (1998) Evoked responses in anaesthesia. British Journal of Anaesthesia ***81:*** *771–781, © The Board of Management and Trustees of the British Journal of Anaesthesia. Reproduced by permission of Oxford University Press/British Journal of Anaesthesia.)*

cervical surgery. Stimuli are administered at about 30 mA and 5 Hz. Recording electrodes may be applied at various points adjacent to the ascending tracts and proximal to the site of surgery. Usually two or more scalp electrodes are used (e.g. one frontal and one cervical), together with a reference and a ground electrode. The recorded responses are passed through a digital signal converter and a band-pass filter (20–1000 Hz). An average response is calculated from as many as 200 individual sweeps and the result displayed on the monitor screen. The period of interest is the first 100 ms after the stimulus, during which a characteristic W-shaped potential is seen in recordings from cortical electrodes (Fig. 17.4). Response latency also depends on the distance between the point of stimulation and the recording electrode.

A baseline SSEP is recorded before induction of anaesthesia. If subsequent intraoperative spinal manipulation completely prevents conduction of the ascending impulse, the SSEP disappears altogether. When conduction is merely impaired, an increase in SSEP latency and a decrease in amplitude are observed. A 50% increase in latency compared to baseline would be regarded as a critical change and the operator would inform the surgeon accordingly. Surgical disruption apart, a number of factors affect SSEPs, including volatile anaesthetic agents, nitrous oxide, hypothermia, hypoxia and hypotension.

Motor-evoked potentials (MEPs)

MEP monitoring is now employed routinely in some spinal surgery centres, as SSEPs may have limited ability to detect ischaemia secondary to anterior spinal artery hypoperfusion. For spinal surgery, transcranial electrical stimulation is most commonly used. Though this carries a risk of injury secondary to induced contraction of the mandibular muscles; in practice serious complications are rarely seen. Alternatives include transcranial magnetic stimulation or direct stimulation of the rostral spinal cord.[19] Multiple-pulse stimulations are performed and the results recorded at subcutaneous or intramuscular needle electrodes in arm and leg muscles. Unlike SSEPs, results can be available within 1 min.

As with SSEPs, the technician can inform the surgeon if MEPs disappear or if the stimulus required to elicit a response exceeds a pre-agreed threshold, indicating potential nerve damage. Spontaneous EMG activity consequent on stretch or compression of nerve roots, which can be detected by MEP recording electrodes, may provide useful information even in the absence of deliberate stimulation.[20]

MONITORING 'DEPTH OF ANAESTHESIA'

For decades, anaesthetists have sought a monitor which might reflect the conscious level of patients undergoing general anaesthesia. A number of different variables, which may relate to consciousness, can be derived from the electrical activity of the cerebral cortex. Ideally,

in variables of this sort, the range of values seen in the conscious state should not overlap with that seen in the unconscious state (i.e. a cut-off value would exist which is 100% sensitive and specific for consciousness). Furthermore, ideally any cut-off value should not be affected by patient physiology or the choice of anaesthetic agent.[21]

At present, the variables derived by available monitors do not meet these ideals. Current monitors generally work on one of two principles:

1. Spontaneous electroencephalography (EEG)
2. Provoked EEG in the form of auditory evoked potentials (AEP).

EEG

General principles, signal processing and artefact rejection

The EEG represents current flow in the cortical extracellular fluid, which is the result of post-synaptic potentials in cortical neurons. It is acquired through scalp electrodes. In anaesthetic monitors, between one and four electrodes are usually used, together with a reference electrode. Impedance should be kept to a minimum (i.e. less than 5 kΩ). EEG voltage is measured as the potential difference between two electrodes.

The complex waveform of the EEG comprises many individual sine waves, whose frequencies lie from zero to approximately 50 Hz (Fig. 17.5). Classically these are grouped into frequency bands (Table 17.1). For the purposes of analysis, the EEG is split into epochs of 1–4 s.

In order to allow processing, the EEG is converted from a smooth continuous analogue signal into a digital one. The fidelity of this conversion depends on the degree of resolution for both voltage and time. The greater the number of bits used, the smaller the change in voltage that can be translated from analogue to digital. EEG monitors usually use 12–16 bits of resolution. The frequency at which the analogue signal is sampled is also important. Too slow a sampling frequency fails to take account of the fastest sine waves and results in 'aliasing', where the digital signal incorrectly identifies a low-frequency waveform (Fig. 17.6). EEG monitors usually sample at frequencies above 250 Hz. Signal processing then enables rejection of artefact due to the electrocardiogram, ocular movement or mains interference.

Time and frequency domain analysis and data presentation

The EEG is then subjected to *time-domain* and *frequency-domain* analysis. The most widely used example of the former is the calculation of a *burst suppression ratio* (BSR). In deeply anaesthetized patients, the EEG may consist of *isoelectric periods* (suppression) interspersed with bursts of normal activity. The BSR is the fraction of time in an epoch in which the EEG is suppressed and is usually averaged over 60 s.

In frequency-domain analysis, each individual epoch is subjected to fast Fourier transformation. This mathematical process breaks down the EEG waveform into

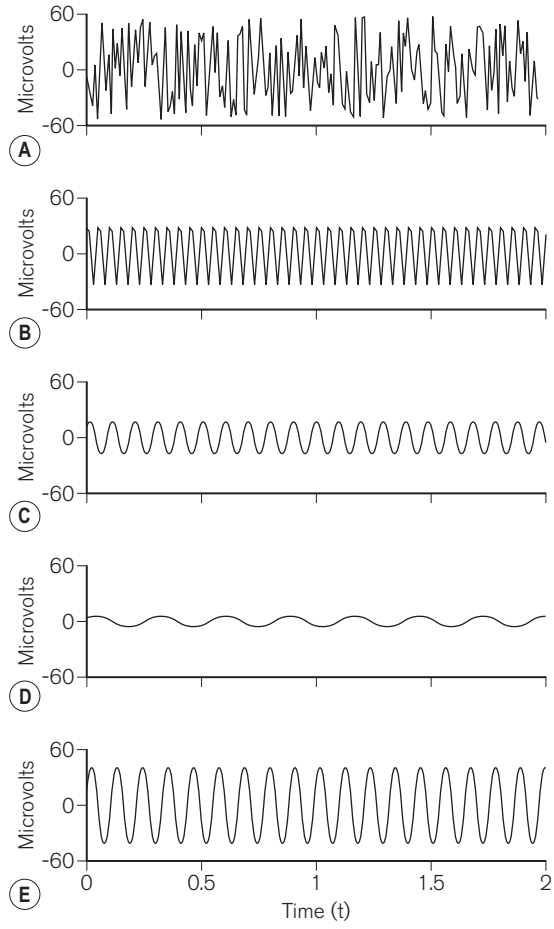


Figure 17.5 **A.** A 2-s epoch from an unprocessed EEG. Also pictured are some of its constituent sine waves, whose respective amplitudes, phase angles and frequencies are **B.** 30 μV, 60° and 20 Hz, **C.** 15 μV, 30° and 10 Hz, **D.** 4 μV, 15° and 4 Hz, **E.** 40 μV, 25 and 9 Hz.

Table 17.1 Electroencephalographic frequency bands

BAND NAME	APPROXIMATE FREQUENCY RANGE (Hz)
Alpha	8–13
Beta	13–30
Theta	4–8
Delta	<4

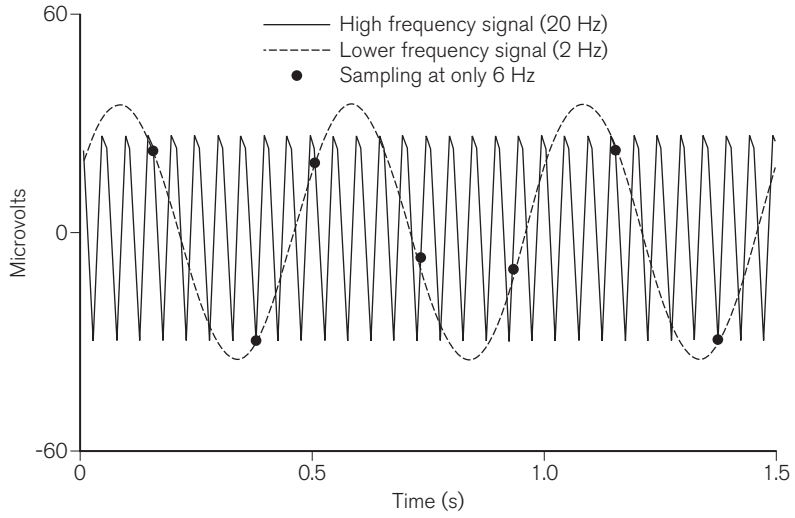


Figure 17.6 Aliasing. A high frequency signal at 20 Hz is incorrectly identified as a lower frequency signal at 2 Hz because the signal is sampled at only 6 Hz.

its constituent sine waves, from which a power (μV^2) vs frequency (Hz) histogram can be derived. On many monitors, the frequency histograms from sequential epochs can be presented as a *compressed spectral array* (CSA) or a *density spectral array* (DSA) (Fig. 17.7).

For the purposes of objective comparison, the frequency spectrum may be represented by a summary variable. Two commonly used variables are the median frequency (MF), which divides the power in the spectrum into two equal halves, and the 95% spectral edge frequency (SEF), below which 95% of the spectral power lies (Fig. 17.8). As anaesthesia deepens, lower EEG frequencies generally predominate and there is a concomitant fall in MF and SEF. Despite this broad relationship, MF and SEF are unsatisfactory measures of consciousness during anaesthesia. They have not been related to clinical endpoints, nor are they independent of anaesthetic agents used.

Bispectral analysis and the EEG bispectral index

The bispectral index (BIS) is an EEG-derived variable which is calculated and displayed by several devices manufactured by Aspect Medical Systems: such as the BIS VISTA monitoring system (Fig. 17.9) and BIS modules in multimodal monitors produced by other manufacturers. *Bispectral analysis* of the EEG signal, a method which is integral to the calculation of BIS, is a method of addressing relationships among signal constituents in the EEG. The use of MF and SEF as indices of anaesthesia described previously relies on assumptions about the linearity of EEG data. In linear systems, there is a simple proportional relationship between cause and effect. However, this situation is demonstrably not the case for many natural phenomena, including the EEG where different frequencies within the signal may not be independent of each other. Bispectral analysis, a higher order analysis based on the phase relationships of individual frequency components to each other, goes some way to addressing this issue and has long been used to examine wave patterns in oceanography. In the EEG, it is particularly applicable to the relationships between the constituent sine waves (described above under frequency domain analysis). These relationships – '*phase coupling*' – have some relevance to the EEG effects of anaesthesia, becoming progressively more pronounced with increasing anaesthetic dose.

BIS is calculated from the EEG using a proprietary algorithm, which incorporates data both from bispectral analysis and more traditional methods. The constituent elements are:

1. two time-domain features:
   a. BSR, see above
   b. QUAZI suppression index reflecting the proportion of near (or quasi) isoelectric activity
2. two frequency-domain features:
   a. 'SynchFastSlow' (from bispectral analysis – the log ratio of bispectral power in the 0.5–47 Hz and 40–47 Hz frequency ranges)
   b. 'BetaRatio' (the log ratio of power in the 30–47 Hz and 11–20 Hz frequency ranges).[22]

The BetaRatio constituent relates to a phenomenon called 'beta activation', a paradoxical increase in relative beta power occurring at low brain concentrations of some anaesthetics. The overall product of the BIS algorithm is a dimensionless scale from 1 to 100, which indicates the likelihood of consciousness. During general anaesthesia,

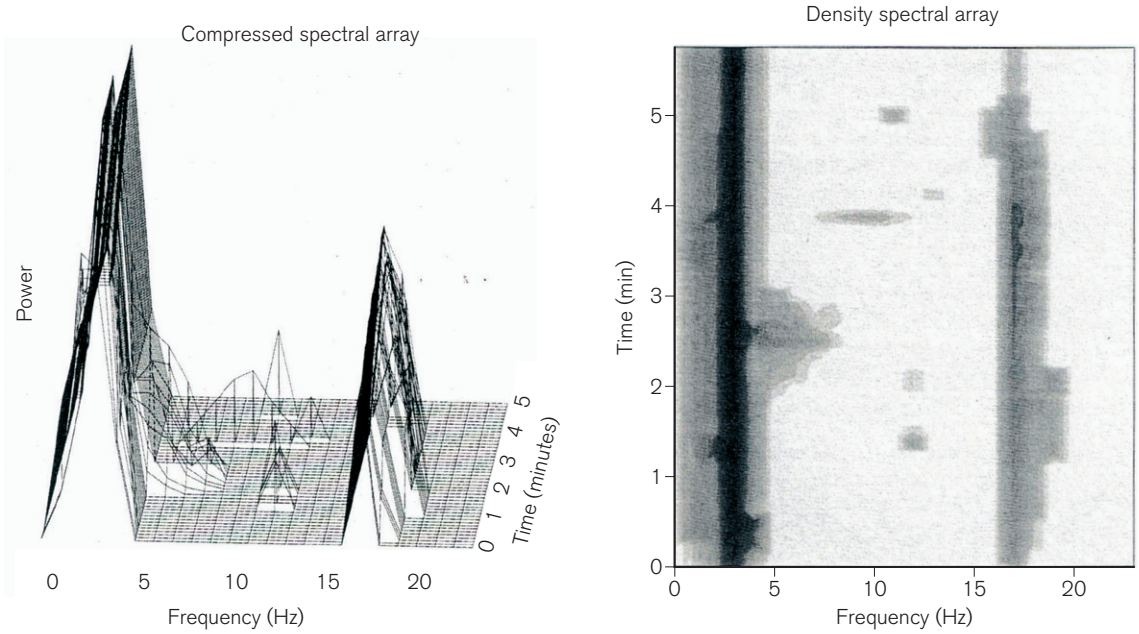


Figure 17.7 Spectral array displays. These are created by fast Fourier transformation of sequential epochs of raw EEG signal. Data from each epoch are transformed to produce a power *versus* frequency histogram. In the compressed spectral array (CSA) above, histograms derived from sequential epochs are plotted in pseudo-three-dimensional fashion. The density spectral array shown is derived from the same EEG data as the CSA. Each histogram is represented in greyscale, with larger values depicted by darker shades. In each case, data from the most recent epoch are added to the bottom of the display.
(*Rampil IJ (1998) A primer for EEG signal processing.* Anesthesiology ***89:*** *980–1002, with permission.)*

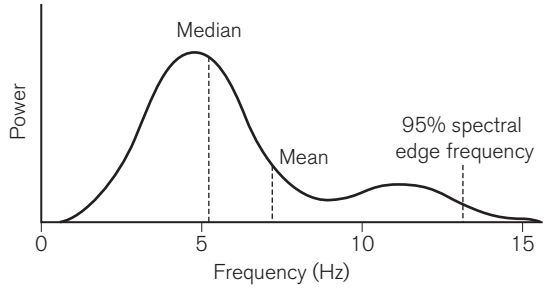


Figure 17.8 The EEG frequency spectrum and derived variables.

a BIS of less than 60 is said to indicate a negligible chance of recall.

The development of the BIS algorithm, and of bispectral index monitoring in general, has been described in several reviews.[22,23,24] The first algorithm, BIS 1.0, was released in 1992. It was constructed using EEG data acquired from young, fit patients under anaesthesia. Originally, the EEG features that best predicted movement on incision were identified and combined in an index which was then tested prospectively. Research in the early 1990s, however, indicated that the neural mechanisms underlying the movement response to incision differed from those concerned with hypnosis. As a result, BIS 2.0 was formulated using the existing database, to predict hypnotic rather than motor endpoints.

By version 3.0 (1995), the database had increased to about 1500 anaesthetics. This version, and subsequent ones, has represented attempts to improve performance of BIS at very deep or light levels of anaesthesia and to enhance removal of electromyographic or electrocautery-related artefact. To this end, the latest VISTA generation of BIS monitors use the 4.0 or 4.1 versions of the algorithm. In these versions BIS is calculated using data from sensors containing four rather than three electrodes, such as the Quatro (Fig. 17.10), Extend, and semi-re-usable sensors (all Aspect Medical Systems), in which the additional electrode is placed above and lateral to the eyebrow. A bilateral sensor has also been released recently, though its advantages are unclear at present.

While BIS is now widely used, its reliability as an indicator of hypnosis during general anaesthesia depends on the agent. It is ineffective during ketamine anaesthesia and doubts remain over the precise relationship between BIS and clinical state during nitrous oxide sedation and high-dose opioid anaesthesia. In recent years, authors have suggested that the BIS algorithm is unnecessarily complex. They argue that a variable, analogous to SynchFastSlow, can be calculated from the EEG without bispectral analysis. This variable, PowerFastSlow, requires fewer data for its calculation. It too measures phase coupling, using a

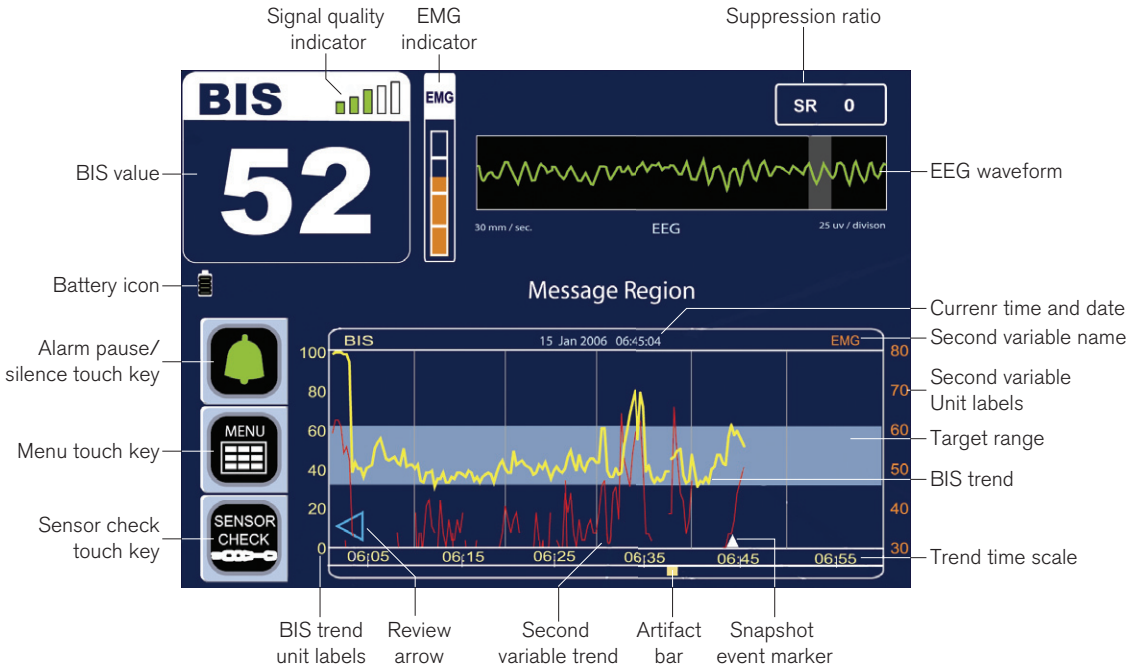


Figure 17.9 Annotated diagram of the display panel: Aspect BIS VISTA monitor.
Image courtesy of Covidien, Ireland.



Figure 17.10 The BIS QUATRO sensor for use with the bispectral index monitor.
Image courtesy of Covidien, Ireland.

phenomenon known as bicoherence. Unlike the bispectrum, this is independent of signal amplitudes. In patients receiving a propofol, alfentanil and isoflurane anaesthetic, PowerFastSlow appears to predict the anaesthetized state as well as SynchFastSlow.[25]

Other spontaneous EEG processing devices

Other monitors which record and process the EEG are available. The SEDLine (Hospira), formerly known as the Patient State Analyzer PSA-4000, was developed in a broadly similar fashion to the BIS monitor, using EEG databases and clinical correlates. Most of the EEG descriptors used in the algorithm for Patient State Index (PSI) calculation, relate to EEG power, though the suppression ratio is also taken into account. Six electrodes are placed on the forehead, to acquire data from four EEG channels. The resulting PSI ranges from 0 to 100 and PSI 25–50 is recommended for surgical anaesthesia.[26]

In the Narcotrend monitor (MonitorTechnik), the EEG is acquired from one reference and two recording electrodes on the forehead. Artefact is rejected, the data are analyzed and an algorithm applied in order to assign a Narcotrend stage. There are six such stages – A (awake) to F (general anaesthesia with increasing burst suppression). A recent version also includes a numerical index. The algorithm is distinct from the BIS and PSI algorithms in that it was developed purely as a means for objective analysis of the EEG waveform, using time and frequency domain information. No clinical correlates were involved.[27]

Effective intraoperative titration of general anaesthetic has been demonstrated using all three monitors described above, and the M-Entropy module (see below), with monitored patients recovering more quickly than controls.

Other commercial monitors based on the spontaneous EEG included the Cerebral State Monitor (Danmeter A/S). This is a small battery-operated handheld device attached by electrodes to the forehead. Both proprietary and routine ECG electrodes can be used. It acquires data from a single

EEG channel. Fuzzy logic (if x and y, then z) is then applied to four calculated EEG parameters, namely alpha ratio, beta ratio, the difference between these ratios and burst suppression ratio, to determine the Cerebral State Index (CSI). The CSI is a dimensionless number scaled from 0 to 100.[28] Danmeter A/S ceased trading in 2008, although the products are still in circulation.

The SNAP II (Everest Biomedical Instruments) is based on a personal digital assistant, onto which can be 'snapped' a monitoring module. The single-channel EEG is acquired from a forehead electrode strip and the SNAP II index is calculated using low (0.1–18 Hz) and high (80–420 Hz) frequency components of the EEG. The index runs from zero to 100, the latter indicating the patient is fully awake. A SNAP index of 50–65 is recommended for general anaesthesia.[29]

Entropy

In 2003, Datex-Ohmeda released a depth of anesthesia module for their S/5 anaesthesia monitor (Fig. 17.11) based on the entropy of the spontaneous EEG. If the awake EEG is characterized by a chaotic signal, then decreasing levels of consciousness are associated with a less disordered signal, as the number of signal generators diminish and slower wave activity becomes more dominant. By calculating the amount of disorder in the power spectrum of the EEG signal, it is suggested that anaesthetic depth may be objectively estimated. In contrast to the often opaque proprietary algorithms above, this system uses a relatively simple mathematical calculation of Shannon entropy with the resultant numerical scale deliberately manipulated to correlate with the now familiar BIS values. A three electrode sensor similar in appearance and application to that for BIS is used to acquire a single-channel EEG. From this, the module calculates two values: one between 0 and 91 for '*state entropy*' (SE), which reflects cortical activity over the frequency range 0.8–32 Hz, and a second between 0 and 100 for '*response entropy*' (RE) over the frequency range 0.8–47 Hz, to include components of the frontalis muscle electromyogram (EMG).



Figure 17.11 Datex–Ohmeda M Entropy module and electrode array.

The manufacturers recommend that both values should be between 40 and 60 during surgery under general anaesthesia. RE becomes equal to SE when the EMG power is zero, otherwise it is always higher than SE and rapid rises in RE are said to reflect relative analgesic inadequacy. Where SE is above 60, a higher anaesthetic dose is required. Where SE is satisfactory, but RE is more than 5–10 units above the SE value, more analgesic is needed. It is hypothesized that, in the case of deficient analgesia in a non-paralyzed patient, EMG facial activity increases before any change in the EEG activity, leading to an increase in RE before any change in SE.

Although global correlation with BIS appears good, agreement is poor. This may reflect differences in scale, despite numerical adjustment. In general, the M-Entropy module appears to perform no better than BIS in terms of correlation with effect site drug concentration and prediction of clinical endpoints. In sedated patients the frontalis EMG renders interpretation of SE and RE values difficult.[30]

Auditory evoked potentials

Loss of consciousness under general anaesthesia is accompanied by changes in the electrical response of the cerebral cortex to an auditory stimulus. The period of particular interest is the early cortical response, illustrated in Fig. 17.12, which occurs approximately 10–100 ms after the stimulus (the mid-latency auditory evoked potential, or MLAEP). MLAEP waves are generated in the medial geniculate and primary auditory cortex. Anaesthesia increases the latency and decreases the amplitude of MLAEP waves. Threshold values for both Na and Pb latencies have been proposed as indicators of unconsciousness during anaesthesia. More complex MLAEP-derived indices, which better reflect its overall morphology, have potential clinical advantages and may feature in future monitors.[31] Particular frequency components within the MLAEP may also have a role.[32]

The A-Line AEP monitor/2 (Danmeter) is a hybrid spontaneous EEG/AEP monitor (Fig. 17.13). For the AEP component 65–70 dB click stimuli are administered at 9 Hz through headphones, the response being recorded with scalp electrodes. A 'Click Detection' function alerts the user if the auditory stimulus ceases for some reason. The response signal undergoes pre-processing, during which artefact is rejected and band-pass filtering applied. Subsequent extraction of the AEP from background cortical activity takes time and results in some latency. In current versions of the monitor, the replacement of 'moving time averaging' with an 'autoregressive model' (using a proprietary mathematical method called ARX modelling) has reduced the update delay time from 35 to 6 s. The latest

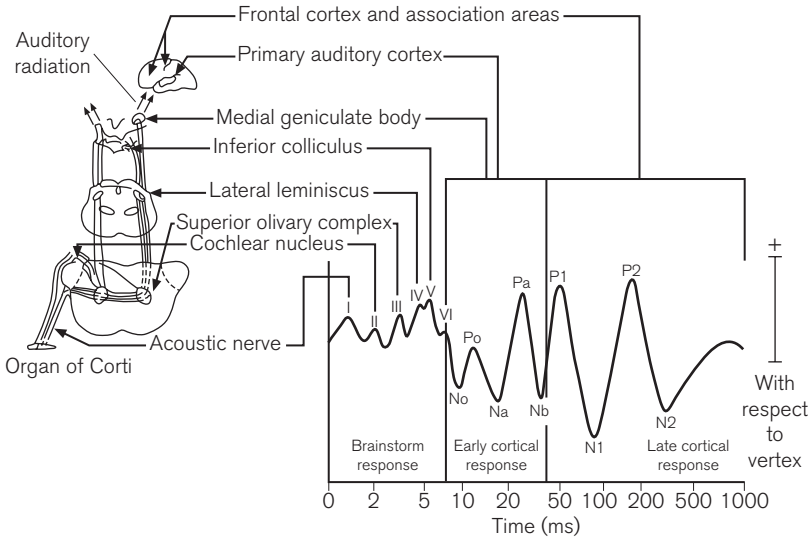


Figure 17.12 The auditory evoked response with its anatomical basis.
(Thornton C, Sharpe RM (1998) Evoked responses in anaesthesia. British Journal of Anaesthesia ***81:*** *771–781, © The Board of Management and Trustees of the British Journal of Anaesthesia. Reproduced by permission of Oxford University Press/British Journal of Anaesthesia.)*

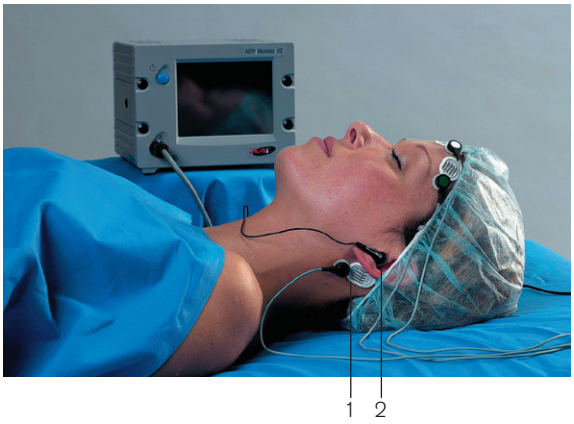


Figure 17.13 The Danmeter A-Line AEP monitor: (1) recording scalp electrodes; (2) headphones.

A-line ARX Index (AAI) is the sum of absolute differences in the 20–80 ms window of the AEP and is a unitless index ranging from 0 to 99. This is preferentially derived from the AEP, but the EEG signal is used if AEP values are weak. This device is no longer in production.

aepEX monitor

Manufactured by Medical Device Management Ltd, this small handheld monitor, which is also available as a plug-in unit for modular monitors, generates a single number between 0 and 100 based on analysis of auditory evoked potentials (Fig. 17.14). The aepEX index, previously called the 'AEP index' and the 'level of arousal score', reflects the morphology of the MLAEP curves. It is calculated as the sum of the square root of the absolute difference between every two successive 0.56 ms segments of the AEP waveform.[33] The AEP curves recorded from scalp electrodes are produced by averaging 256 sweeps of 144 ms following the presentation of 7 Hz auditory clicks of 1 ms duration at 70 dB above normal hearing threshold. Stimuli are administered through small earphones with single-use gels, available in three sizes. The EEG signal is detected by three disposable sensors: two on the forehead and one on the mastoid process. By using a moving time averaging technique results are updated at 1 s intervals. Typical values in a conscious patient are 65–85 and, during surgery, 30–45.

The aepEX is said by the manufacturers to work equally well with all anaesthetic agents including ketamine and nitrous oxide. There is some evidence to support this particularly where ketamine is given in association with remifentanil.[34,35]

Clinical use of depth of anaesthesia monitoring

Of the above technologies, at present BIS has European and US market dominance and the largest volume of supporting evidence. AEP index and Entropy also appear to show efficacy in attempting to detect unconsciousness. In a direct experimental comparison during propofol sedation, an AEP-derived variable (the AEP index) was better



Figure 17.14 The aepEX monitor together with sensors and earphones.
Photograph courtesy of Medical Device Management Ltd. UK.

than BIS at distinguishing the conscious from the unconscious state.[33] However, results from a large-scale trial indicate that BIS monitoring can reduce the incidence of intraoperative awareness in patients at risk of this complication,[36] although a trial comparing a BIS based protocol and one based on the measurement of end tidal anaesthetic agent showed no advantage.[37] At present, there is no corresponding evidence for the effectiveness of AAI-guided general anaesthesia and Danmeter, the manufacturer of both AEP/2 and the Cerebral State Monitor, has recently ceased trading.

New ways of using the information in the spontaneous EEG to reflect the hypnotic effect of anaesthetic agents appear frequently in the literature. In general, the drive is to develop a less complicated algorithm than BIS, while retaining its clinical utility. On most occasions, the reliability of the new technology is assessed in terms of its agreement with simultaneous BIS readings. In the more rigorous studies, the ability of new technology to predict clinical endpoints is compared with that of BIS.

Limitations

Non-anaesthetic factors affecting cerebral metabolism, e.g. hypotension, hypoxaemia, hypothermia and hypoglycaemia, can affect EEG-derived indices of 'depth of anaesthesia'. The precise contribution of these factors to an anaesthetic effect, as compared to the output of the monitors, is not yet elucidated. Additionally, opiates, especially at high dose, have a particular effect on the EEG whilst being devoid of the ability to prevent recall. Other drugs: psychotherapeutic agents, ketamine and nitrous oxide may introduce further difficulties in producing inconsistent effects on the measured entropy of the EEG or producing an anaesthetic effect not legible to BIS.

It is important in using all monitors of depth of anaesthesia that we do not conflate the different questions that may be asked:

- Is the patient asleep or awake at present?
- How much effective anaesthetic agent is active in the patient?
- What will be the effect of surgical stimulation on arousal?
- Will there be movement in response to stimulation?
- Is there or will there be information processing or retention taking place?

The EEG analysis technique used by a depth of anaesthesia monitor, the patient physiological state and the particular anaesthetic agents in use will govern the veracity with which each of these questions may be answered.

It is conceivable that in due course we will decide that spontaneous EEG-based monitors in assessing current cortical activity will always tend to lack the predictive potential of monitors based on provoked EEG, which by their nature are assessing the ability of the brain to respond to a stimulus.

PAIN MEASUREMENT

In an attempt to facilitate detection of pain in patients unable to communicate reliably, a skin conductance algesimeter, the MEDSTORM AS 2005 monitor (Med-Storm Innovation), has recently been developed. Nociceptive stimuli increase sympathetic stimulation to skin sweat glands, thereby raising electrical conductance. The monitor acquires data from three palmar electrodes. The number of fluctuations per second in skin conductance (NFSC) appears to be a more useful variable than mean skin conductance, which is highly dependent on electrode position and type.

In one prospective study of the monitor in postoperative adult patients, NFSC correlated with pain score. Using a predetermined cut-off value for NFSC of 0.1, the monitor distinguished between those with no/mild pain and those with moderate/severe pain with a sensitivity and specificity of 88.5% and 67.7% respectively.[38] Though such a monitor might be helpful in some circumstances, the skin conductance method appears to be affected by anticholinergic agents.[39]

Though changes in Vmax are widely regarded as being more significant than absolute values, it is unclear what constitutes a critical reduction. In one large study, ischaemia after clamping was considered severe if Vmax fell to 0–15% of baseline, mild if 16–40% and absent if greater than 40%.[41] A significant decrease on clamping may influence the surgical decision to shunt. Subsequent Vmax changes may allow detection of shunt occlusion and intraoperative emboli, the latter producing a characteristic sound.

As well as decreased CBF during surgery, TCD allows detection of increased flow postoperatively in hyperperfusion syndrome. This complication occurs in about 1% of patients undergoing CEA. Prompt recognition and administration of appropriate therapy may reduce the risk of cerebral haemorrhage.

Though TCD monitoring has been attempted in cerebral aneurysm surgery and resection of intracranial arteriovenous malformations, practical difficulties limit its usefulness.

SSEPs

Cortical SSEPs are affected by large reductions in CBF. Median nerves may be stimulated bilaterally and evoked potentials recorded at scalp electrodes as described above. A baseline response is recorded before clamping and again about a minute after. Subsequent recording at regular intervals is recommended as delayed changes are occasionally seen. SSEPs have been used successfully as a basis for selective shunting in CEA.[42] On carotid clamping, a decrease in N20/P25 amplitude of greater than 50% from baseline has been proposed as an indicator of cortical dysfunction.[43]

EEG

Cerebral ischaemia occurring on carotid clamping may induce changes in ipsilateral EEG frequencies and amplitudes. Raw EEG monitoring during carotid endarterectomy (CEA) has been advocated, although data interpretation requires considerable expertise. Processed EEG monitors may be more practical, but BIS appears unreliable for detection of cortical ischaemia[44] and the predictive sensitivity of intraoperative EEG monitoring for immediate postoperative neurological deficit is poor.[45]

Jugular bulb oximetry

More usually used in the intensive care unit, jugular bulb oximetry also allows intraoperative assessment of cerebral oxygen extraction during cerebral aneurysm, tumour and haematoma surgery.[46] An oximetric catheter, e.g. Opticath (Abbott), is introduced into the internal jugular vein cranially until the jugular venous bulb is encountered. Jugular venous oxygen saturation ($SjvO_2$) reflects the oxygen '*supply:demand*' ratio and is usually 60–70%. Normal values give limited reassurance as they do not exclude focal cerebral ischaemia. An $SjvO_2$ of 90% or more is seen in hyperaemia – at less than 50%, $SjvO_2$ indicates increased oxygen extraction and impending ischaemia.

Cerebral oximetry

Near infrared spectrometry (e.g. the Somanetics Invos Cerebral Oximeter, Covidien, Ireland) is a non-invasive method of continuously monitoring cerebral oxygen saturation. Regional cerebral oxygen saturation (rsO2) is measured to reflect cerebral perfusion. While the technique may have prognostic potential for cerebral function after cardiac surgery, it appears insufficiently sensitive and specific to predict the need for shunting during carotid endarterectomy.[47]

REFERENCES

1. Saitoh Y, Kaneda K, Hattori H, Nakajima H, Murakawa M. Monitoring of neuromuscular block after administration of vecuronium in patients with diabetes mellitus. *Br J Anaesth* 2003;**90**:480–6.
2. Brull SJ, Silverman DG. Pulse width, stimulus intensity, electrode placement, and polarity during assessment of neuromuscular block. *Anesthesiology* 1995;**83**:702–9.
3. Claudius C, Viby-Mogensen J. Acceleromyography for use in scientific and clinical practice: a systematic review of the evidence. *Anesthesiology* 2008;**108**:1117–40.
4. Trager G, Michaud G, Deschamps S, Hemmerling TM. Comparison of phonomyography, kinemyography and mechanomyography for neuromuscular monitoring. *Can J Anaesth* 2006;**53**:130–5.
5. Dahaba AA, Rehak PH, List WF. Assessment of accelerography with the TOF-GUARD: a comparison with electromyography. *Eur J Anaesthesiol* 1997;**14**(6):623–9.
6. Suzuki T, Fukano N, Kitajima O, Saeki S, Ogawa S. Normalization of acceleromyographic train-of-four ratio by baseline value for detecting residual neuromuscular block. *Br J Anaesth* 2006;**96**:44–7.
7. Samet A, Capron F, Alla F, Meistelman C, Fuchs-Buder T. Single acceleromyographic train-of-four, 100-Hertz tetanus or double-burst stimulation: which test performs better to detect residual paralysis? *Anesthesiology* 2005;**102**:51–6.
8. Tsui BC, Wagner AM, Cunningham K, Perry S, Desai S, Seal R. Can continuous low current electrical stimulation distinguish insulated needle position in the epidural and intrathecal spaces in pediatric patients? *Paediatr Anaesth* 2005;**15**:959–63.

9. Kaiser H, Niesel HC, Klimpel L, Bodenmueller M. Prilocaine in lumbosacral plexus block–general efficacy and comparison of nerve stimulation amplitude. *Acta Anaesthesiology Scandinavica* 1992;**36**:692–7.
10. Gurnaney H, Ganesh A, Cucchiaro G. The relationship between current intensity for nerve stimulation and success of peripheral nerve blocks performed in pediatric patients under general anesthesia. *Anesth Analg* 2007;**105**:1605–9, table.
11. Szerb J, Persaud D. Long current impulses may be required for nerve stimulation in patients with ischemic pain. *Can J Anaesth* 2005;**52**:963–6.
12. Hadzic A, Vloka J, Hadzic N, Thys DM, Santos AC. Nerve stimulators used for peripheral nerve blocks vary in their electrical characteristics. *Anesthesiology* 2003;**98**:969–74.
13. Tsui BC, Pillay JJ, Chu KT, Dillane D. Electrical impedance to distinguish intraneural from extraneural needle placement in porcine nerves during direct exposure and ultrasound guidance. *Anesthesiology* 2008;**109**:479–83.
14. Cory PC. Increased impedance on nerve stimulator display may actually reflect a decrease in total system impedance. *Anesthesiology* 2009;**110**:1192–4; author reply 1194–5.
15. Engelhardt L, Grosse J, Birnbaum J, Volk T. Inhibition of a pacemaker during nerve stimulation for regional anaesthesia. *Anaesthesia* 2007;**62**:1071–4.
16. Rozner MA. Peripheral nerve stimulators can inhibit monitor display of pacemaker pulses. *J Clin Anesth* 2004;**16**:117–20.
17. Snyder SK, Hendricks JC. Intraoperative neurophysiology testing of the recurrent laryngeal nerve: plaudits and pitfalls. *Surgery* 2005;**138**:1183–91.
18. Nuwer MR, Dawson EG, Carlson LG, Kanim LE, Sherman JE. Somatosensory evoked potential spinal cord monitoring reduces neurologic deficits after scoliosis surgery: results of a large multicenter survey. *Electroencephalogr Clin Neurophysiol* 1995;**96**:6–11.
19. Pajewski TN, Arlet V, Phillips LH. Current approach on spinal cord monitoring: the point of view of the neurologist, the anesthesiologist and the spine surgeon. *Eur Spine J* 2007;**16**(Suppl 2):S11529.
20. Owen JH, Kostuik JP, Gornet M, Petr M, Skelly J, Smoes C, et al. The use of mechanically elicited electromyograms to protect nerve roots during surgery for spinal degeneration. *Spine* 1994;**19**: 1704–10.
21. Drummond JC. Monitoring depth of anesthesia: with emphasis on the application of the bispectral index and the middle latency auditory evoked response to the prevention of recall. *Anesthesiology* 2000;**93**: 876–82.
22. Rampil IJ. A primer for EEG signal processing in anesthesia. *Anesthesiology* 1998;**89**:980–1002.
23. Johansen JW, Sebel PS. Development and clinical application of electroencephalographic bispectrum monitoring. *Anesthesiology* 2000;**93**:1336–44.
24. Johansen JW. Update on bispectral index monitoring. *Best Pract Res Clin Anaesthesiol* 2006;**20**:81–99.
25. Miller A, Sleigh JW, Barnard J, Steyn-Ross DA. Does bispectral analysis of the electroencephalogram add anything but complexity? *Br J Anaesth* 2004;**92**:8–13.
26. Drover D, Ortega HR. Patient state index. *Best Pract Res Clin Anaesthesiol* 2006;**20**:121–8.
27. Kreuer S, Wilhelm W. The Narcotrend monitor. *Best Pract Res Clin Anaesthesiol* 2006;**20**:111–19.
28. Jensen EW, Litvan H, Revuelta M, Rodriguez BE, Caminal P, Martinez P, et al. Cerebral state index during propofol anesthesia: a comparison with the bispectral index and the A-line ARX index. *Anesthesiology* 2006;**105**:28–36.
29. Bischoff P, Schmidt G. Monitoring methods: SNAP. *Best Pract Res Clin Anaesthesiol* 2006;**20**:141–6.
30. Bein B. Entropy. *Best Pract Res Clin Anaesthesiol* 2006;**20**:101–9.
31. Kumar A, Anand S, Yaddanapudi LN. Comparison of auditory evoked potential parameters for predicting clinically anaesthetized state. *Acta Anaesthesiology Scandinavia* 2006;**50**:1139–44.
32. Scheller B, Schneider G, Daunderer M, Kochs EF, Zwissler B. High-frequency components of auditory evoked potentials are detected in responsive but not in unconscious patients. *Anesthesiology* 2005;**103**: 944–50.
33. Gajraj RJ, Doi M, Mantzaridis H, Kenny GN. Comparison of bispectral EEG analysis and auditory evoked potentials for monitoring depth of anaesthesia during propofol anaesthesia. *Br J Anaesth* 1999;**82**:672–8.
34. Kang H, Lee HK. Changes of bispectral index and auditory evoked potential index during general anesthesia with propofol, midazolam and ketamine. *Korean J Anesthesiol* 2005;**49**: 757–64.
35. Kang H, Scott N, Sutcliffe N, Kenny GNC. *The effects of intravenous ketamine and propofol on the BIS and Aepex*. Archives of European Society for Intravenous Anaesthesia, Annual Scientific Meeting. Lisbon 2004. www.eurosiva.org
36. Myles PS, Leslie K, McNeil J, Forbes A, Chan MT. Bispectral index monitoring to prevent awareness during anaesthesia: the B-Aware randomised controlled trial. *Lancet* 2004;**363**:1757–63.
37. Avidan MS, Zhanjg L, Burnside BA, Finkel KJ, Searleman AC, Selvidge JA, et al. Anaesthesia awareness and the bispectral index. *NEJM* 2008;**358**:1097–108.
38. Ledowski T, Bromilow J, Wu J, Paech MJ, Storm H, Schug SA. The assessment of postoperative pain by monitoring skin conductance: results of a prospective study4. *Anaesthesia* 2007;**62**:989–93.
39. Ledowski T, Preuss J, Schug SA. The effects of neostigmine and glycopyrrolate on skin conductance as a measure of pain. *Eur J Anaesthesiol* 2009;**26**:777–81.
40. Jacob T, Hingorani A, Ascher E. Carotid Artery Stump Pressure (CASP) in 1135 consecutive endarterectomies under general anesthesia: an old method that survived the test of times. *J Cardiovasc Surg (Torino)* 2007;**48**:677–81.

41. Halsey JH Jr. Risks and benefits of shunting in carotid endarterectomy. The International Transcranial Doppler Collaborators. *Eur J Anaesthesiol* 1992;**23**:1583–7.
42. Schwartz ML, Panetta TF, Kaplan BJ, Legatt AD, Suggs WD, Wengerter KR, et al. Somatosensory evoked potential monitoring during carotid surgery. *Cardiovasc Surg* 1996;**4**:77–80.
43. Stejskal L, Kramar F, Ostry S, Benes V, Mohapl M, Limberk B. Experience of 500 cases of neurophysiological monitoring in carotid endarterectomy. *Acta Neurochir (Wien)* 2007;**149**: 681–8.
44. Culp WC Jr, Kinsky MP, Rios JA, Mendelow AD. Bispectral index monitoring may not reliably indicate cerebral ischaemia during awake carotid endarterectomy. *Br J Anaesth* 2005;**95**:559–60.
45. McCarthy WJ, Park AE, Koushanpour E, Pearce WH, Yao JS. Carotid endarterectomy. Lessons from intraoperative monitoring – a decade of experience. *Ann Surg* 1996;**224**:297–305.
46. Gunn HC, Matta BF, Lam AM, Mayberg TS. Accuracy of continuous jugular bulb venous oximetry during intracranial surgery. *J Neurosurg Anesthesiol* 1995;**7**:174–7.
47. Rigamonti A, Scandroglio M, Minicucci F, Magrin S, Carozzo A, Casati A. A clinical evaluation of near-infrared cerebral oximetry in the awake patient to monitor cerebral perfusion during carotid endarterectomy. *J Clin Anesth* 2005;**17**:426–30.

FURTHER READING

Sleigh JW, Barnard JPM. Entropy is blind to nitrous oxide. Can we see why? *Br J Anaesth* 2004;**92**(2): 159–61.

Bruhn J, Myles PS, Sneyd R, Struys MMRF. Depth of anaesthesia monitoring: what's available, what's validated and what's next? *Br J Anaesth* 2006;**97**(1):85–94.

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Chapter 18

Atmospheric pollution

Sarah Bailey

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Gaseous and volatile agents are widely used in anaesthesia, with an excellent safety profile at an individual level. What is less clear is the effect of long-term exposure to sub-therapeutic levels on the health of those working in environments where there are anaesthetic agents in the ambient atmosphere, and the effect these have when exhaled into the external environment in general. This has led to the development and refinement of equipment to ensure minimal concentrations are present in the direct theatre environment with the aim of protecting those working in these areas.

EFFECTS ON THE ENVIRONMENT

Anaesthetic gasses and vapours are known to have an effect on both ozone depletion and climate change.

The stratosphere is the region of the atmosphere from about 10 to 50 km above the Earth's surface, where ozone plays a vital role absorbing harmful short wavelength ultraviolet radiation from the sun and protecting the earth (Fig. 18.1). Stratospheric ozone is depleted by human-made chemicals including hydrochlorofluorocarbons (e.g. halothane, enflurane, isoflurane) and nitrous oxide.

The contribution of these agents to ozone depletion is a function of their lifetimes in the atmosphere, and these lifetimes depend on the reaction of the drugs with hydroxyl radicals in the troposphere[1] (Table 18.1). The relatively short lifetimes of these agents along with their minimal production means they have been seen as relatively 'ozone friendly'. However, with the reduction of chlorofluorocarbons globally, the influence on ozone depletion by volatile anaesthetics is potentially of increasing importance.[4]

Nitrous oxide, however, has a much longer lifetime (similar to the traditional chlorofluorocarbons, well known for their ozone depleting effects, and which have been successfully reduced since the 1987 Montreal Treaty) and has now been shown to be the single most important ozone-depleting emission. This is expected to remain throughout the 21st century.[5]

The term 'greenhouse effect' was first used in the 1800s to describe the naturally occurring function of gasses in the atmosphere to absorb heat. These gasses (mostly water vapour, carbon dioxide, methane, nitrous oxide and fluorocarbons) absorb and then re-emit longer infrared wavelength energy, warming the Earth to 30°C warmer than it would otherwise be and, therefore, enabling life. It is now a term which has negative connotations and is associated with global warming. This is due to the dramatic increase in fossil fuel combustion in the last century and, with this, carbon dioxide production.

Nitrous oxide is an exceptionally potent greenhouse gas with approximately 290 times greater global warming potential than carbon dioxide. Although the amount of nitrous oxide generated from medical sources compared

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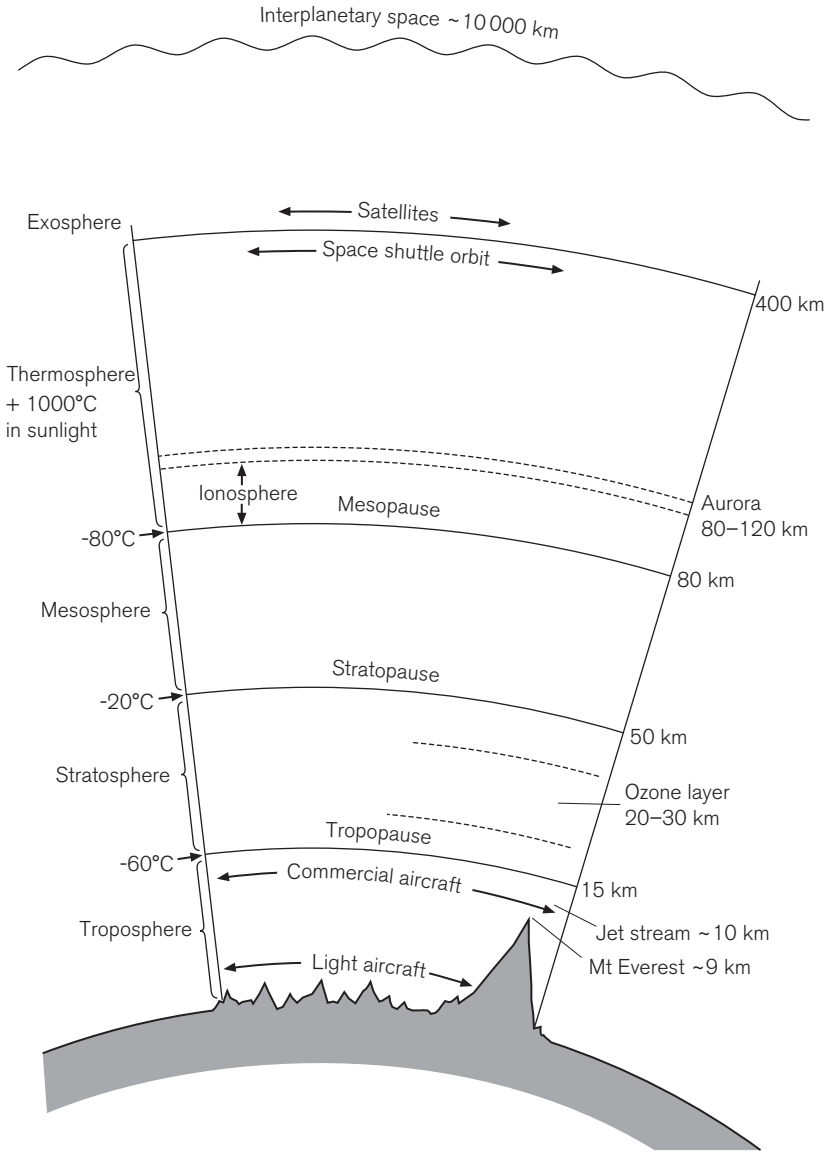


Figure 18.1 The Earth's blanket of air reaching approximately 10 000 km above its surface. Approximate temperatures are also shown.

with the total global production is small, its contribution in the light of environmental issues and pressures is still significant and difficult to ignore. It would, therefore, seem prudent to reduce the use of nitrous oxide when there are alternative agents and techniques available, including the more potent volatile agents, and regional and intravenous anaesthetic techniques.

In anaesthesia, the increasing use of single-use devices in the past decade has also contributed to the global picture of increased carbon dioxide emissions. Within the wider context of atmospheric pollution, recycling in healthcare needs to be addressed. Healthcare institutions lag behind other industries on this due to understandable concerns regarding the potential for contamination from the reuse or recycling of biohazardous material. However, the vast quantities of waste generated in healthcare environments that are not currently recycled will become an issue in the future.

Table 18.1 The effects of anaesthetic gasses on the ozone layer and greenhouse warming

COMPOUND	LIFETIME IN ATMOSPHERE (YEARS)	OZONE DEPLETION POTENTIAL	GREENHOUSE WARMING POTENTIAL	GLOBAL PRODUCTION (TONNES/YEAR)
CFC-11	76	1.00	0.390	350 000
CFC-12	140	1.00	1.000	400 000
Halothane	2	0.36	0.004	1000
Enflurane	6	0.02	0.040	220
Isoflurane	5	0.01	0.030	800
Sevoflurane	1.4	0.00	0.0005	–
*Nitrous oxide	120	0.15	0.040	–

The potential ozone depletion efficacy and greenhouse warming effect are normalized to the principle CFC-12. (Halsey 1996, with permission of The Medicine Group (Education) Ltd[2] (based on original data from 1989)[1].)
*Represents data from the World Meteorological Organization.[3]

EFFECTS ON INDIVIDUALS

Chronic exposure to low concentrations of anaesthetic gasses has been associated with adverse health effects. There have been studies and case reports of these effects since the 1960s, although the evidence is sometimes conflicting. Some animal and human studies[6–11] have suggested that as a result of chronic exposure to inhalational agents amongst theatre personnel there is a demonstrable increase in:

- spontaneous abortion
- fertility problems
- minor congenital abnormalities
- subjective complaints (e.g. headaches, fatigue and nervousness)
- problems with balance control
- cancer (leukaemia and lymphoma)
- effects on immune system (secondary to neutrophil apoptosis)
- liver and renal disease.

In other studies, long-term exposure to nitrous oxide has been shown to result in:

- reduced fertility and increased miscarriage rate in female dental assistants
- litters that are reduced in number and size compared with control animals (rats)
- neurological symptoms indistinguishable from those caused by vitamin B12 deficiency.

The Health and Safety Commission's Advisory Committee on Toxic Substances reviewed the literature on the toxic effects of anaesthetic agents in the workplace in 1996.[12] They made the following conclusions based on the data available:

- There was no evidence in humans that exposure to nitrous oxide or volatile agents (halothane, isoflurane and enflurane) caused developmental defects in the foetus or other reproductive health effects.
- Animals (rats) continuously exposed to high concentrations of nitrous oxide (1000 ppm for over 8 h) demonstrated developmental toxicity to the embryo/foetus, possibly by the inhibition of cell production by nitrous oxide. However, no adverse effects were seen when animals were exposed to nitrous oxide at lower concentrations (500 ppm).
- Pregnant animals exposed to high concentrations of halothane and isoflurane (1000 ppm) showed effects on the development of the foetus. However, there was no convincing evidence when the concentrations of repeated exposure were lower (100 ppm for halothane and 600 ppm for isoflurane).
- There was no evidence from animal studies that suggested enflurane had any adverse effect on the foetus. However, liver damage was demonstrated in mice when exposed to enflurane continuously (>700 ppm).

They therefore set maximum exposure limits for volatiles at levels at which no adverse effects were seen in animal studies and thus represent levels at which there is no evidence to suggest the development of adverse effects in humans.

LEGISLATION

Various organizations in different parts of the developed world have introduced recommendations for maximum

Table 18.2 Exposure limits to anaesthetic gasses and vapours in parts per million (ppm) or as milligrams per cubic metre (mg m^{-3}) and expressed as 8-hour time-weighted averages (8-h TWA)

COUNTRY		N_2O		HALOTHANE		ENFLURANE		ISOFLURANE	
		ppm	mg m^{-3}	ppm	mg m^{-3}	ppm	mg m^{-3}	ppm	mg m^{-3}
UK	2005	100	183	10	82	50	383	50	383
USA	1994	50		50		75 (ACIGH figure, see text)			
Switzerland	1994	100		5		10		10	
Sweden	1994	100		5		10		10	
Italy	1994	50		2		2		2	

acceptable levels of pollution to protect staff working in these areas. Due to the rather inconclusive evidence on adverse effects of volatile agents, these limits vary in different countries.

In the USA, for example, several organizations such as the federal Occupational Safety and Health Administration (OSHA), the National Institute of Occupational Safety (NIOSH) and the American Conference of Industrial Hygienists (ACIGH), have this responsibility. In the UK, it is now the responsibility of the Health and Safety Executive (HSE) following its merger in 2008 with the Health and Safety Commission. In Europe it is the Scientific Committee on Occupational Exposure Limits (SCOEL) that sets these limits under the Chemical Agents Directive.

In the UK, it is a legal requirement that employers control industrial and medical pollution. The legislation takes the form of a government approved code of practice entitled 'Control of Substances Hazardous to Health' (COSHH).[13] This was first introduced in 1988, updated in 1994 and amended annually until 2002. There was a further new edition in 2005 (reprinted in 2008) and further amendments made in accordance with the European Commission's new limits. The HSE's Advisory Committee on Toxic Substances has drawn up this code of practice under Section 16 of the Health and Safety at Work Act (1974), for the purpose of providing practical guidance on the control of substances hazardous to health in the workplace.

It was in 1996 that COSHH defined the safe maximum exposure limits for a wide variety of substances, including anaesthetic gasses and vapours (EH40/96).[14] Since 2005 '*workplace exposure limits*' (WELs) have been the defined limits used to protect workers, replacing the previously used '*maximum exposure limits*' (MELs) and '*occupational exposure standards*' (OES). WELs are defined at concentrations of hazardous substance in the air, averaged over a specified period of time referred to as a time-weighted average (TWA). An 8 h time period is used.

COSHH recommends that: 'Exposure should be controlled to a level to which nearly all the population can be exposed day after day without adverse effect on health'. Recommended exposure limits for anaesthetic gasses and vapours in some countries are set out in Table 18.2. In the UK in 2010 no WELs are available yet for sevoflurane and desflurane. Both OSHA and NIOSH recommend a global ceiling limit (concentrations that must never be exceeded during any part of the day) of 2 ppm for all volatile agents, though they have no regulatory authority. As a rough guide, substances with exposure limits below 100 ppm are considered highly toxic by inhalation, those substances with exposure limits of 100–500 ppm are considered moderately toxic by inhalation and those substances with exposure limits greater than 500 ppm are slightly toxic by inhalation.

The WELs for anaesthetic gasses (enflurane, isoflurane, halothane and nitrous oxide) have remained unchanged since 1996.

There are eight principles of good practice for the control of exposure to substances hazardous to health, published by the Health and Safety Executive in 2005.[15] They are as follows:

1. Design and operate processes and activities to minimize emission, release and spread of substances hazardous to health.
2. Take into account all relevant routes of exposure – inhalation, skin absorption and ingestion – when developing control measures.
3. Control exposure using measures that are proportionate to the health risk.
4. Choose the most effective and reliable control options which minimize the escape and spread of substances hazardous to health.
5. Where adequate control of exposure cannot be achieved by other means, provide, in combination

with other control measures, suitable personal protective equipment.
6. Check and review regularly all elements of control measures for their continuing effectiveness.
7. Inform and train all employees on the hazards and risks from the substances with which they work and the use of control measures developed to minimize the risks.
8. Ensure that the introduction of control measures does not increase the overall risk to health and safety.

To properly protect employees, employers have to follow these principles, and by doing so exposure should be below any WEL.

CONTROL OF POLLUTION

When no steps are taken to avoid pollution, the exposure limits may be exceeded. One study from a 20 hospital survey reported that the levels of halothane varied between 0.1 and 60 ppm (mean of 2.8 ppm) and for nitrous oxide between 10 and 3000 ppm (mean of 388.5 ppm) when scavenging systems were not used.[16] In the same study the installation of an active scavenging system in one particular hospital reduced the anaesthetist's exposure to nitrous oxide (and halothane) from a mean value of 411 ppm (and 1.9 ppm) to a mean value of 24.5 ppm (and <0.1 ppm).

The control of pollution should be tackled using the guidelines recommended in the COSHH in the UK and NIOSH in the USA,[17] namely:

- instilling awareness in personnel working in the potentially affected environment
- installation of effective scavenging equipment (see below)
- ensuring good working practices by:
  - always using the devices provided
  - daily inspection of these devices to ensure that they are functioning
  - considering the use of low-flow systems where appropriate
  - checking for leaks in the breathing system
  - flushing out the breathing system (including the reservoir bag) through the scavenging device provided, at the end of an anaesthetic
  - considering capping off the breathing system at the end of an anaesthetic so as to prevent anaesthetic vapours that have impregnated the breathing hoses from polluting the environment
  - the filling of anaesthetic vaporizers in a fume cupboard that includes a spill tray
  - amending workplace practice by reviewing rotas so that the same personnel are not always working in those areas of highest pollution
- maintaining efficient room air-conditioning so as to remove any pollutant that may have inadvertently escaped (a minimum of 15 changes per hour with a balanced supply and extraction process)
- regularly monitoring the theatre environment.

That which constitutes regular monitoring appears to be the most difficult issue to resolve. Monthly or fortnightly checks might miss a week in which the levels could, due to a fault, contravene COSHH/NIOSH guidelines. An employer (the hospital), if sued by an employee, could well find this case difficult to defend.

THE EXTENT OF POLLUTION

This depends on five factors:

1. the quantity of anaesthetic gasses and vapours employed
2. the employment of a scavenging system and its efficiency
3. the amount of leakage from the anaesthetic equipment
4. the efficiency of the air-conditioning and ventilation system in an operating theatre or anaesthetic room
5. the size and layout of the operating theatre and any other place where anaesthetic vapours are used.

Anaesthetic gasses and vapours

The quantity of gasses and vapours used may vary considerably, depending on the breathing system used. At one extreme is the Mapleson D system, where there may be a fresh gas flow of about 8 L min^{-1}, of which 70% may be nitrous oxide, and to which other volatile anaesthetic agents may be added. At the other extreme is the low-flow circle system, where flows may be reduced to less than 1 L min^{-1}. Also, there is substantial pollution from unscavenged Entonox demand valves used in maternity units.

The employment of a scavenging system and its efficiency

Surplus anaesthetic gasses and vapours are vented from a breathing system or ventilator, via an expiratory valve and, if allowed to escape at this point, would pollute the immediate environment. The valve is normally adapted to discharge into a scavenging system, which collects the escaping gas and vents it to the atmosphere remote from populated areas. The efficiency of the scavenging system depends on its rate of extraction and the gas-tight fit of its components. The former must be greater than the discharge of pollutant gasses, in order to be effective.

These systems are discussed in greater detail later in the chapter.

Leakage

However efficient a scavenging system may be, its purpose will be defeated if gasses and vapours are permitted to escape from the apparatus. Overt leaks from the high-pressure and regulated-pressure parts of the anaesthetic machine may be easily detected. Leaks from the breathing system may be less obvious, however, and may even be due to diffusion through the rubber or neoprene parts. The latter often absorb significant quantities of some of the volatile agents during the administration of one anaesthetic, only to release them during the next anaesthetic. For this reason, new and unused breathing attachments should be used for the administration of anaesthesia to a patient who exhibits sensitivity to a particular anaesthetic agent, for instance in the case of malignant hyperpyrexia.

Leakage may also result from carelessness when vaporizers are refilled. It has been advocated that refilling of vaporizers should take place in a fume hood.

The efficiency of the air-conditioning system

The frequency of air changes is often quoted as a measure of the efficiency of an air-conditioning system. A figure of 20 changes per hour is usually considered satisfactory. However, the circulation of air throughout the theatre is often uneven. The recovery area, where the patient exhales anaesthetic agents, is often poorly ventilated and there are seldom arrangements for scavenging. The nurse attending the patient is often in direct line with the exhaled gasses.

There are two further considerations:

1. Some air-conditioning systems are wholly or partially recirculating, and may result in the vapours from one location polluting another.
2. Thought must be given to the siting of the external outlet of the extraction system, which again may pollute other areas in which people work.

The size of the premises

'Dental chair' anaesthetics (where these are still permitted) for dental treatment are often administered in small rooms. Such practices are rare now in the UK where only a few dental surgeries offer sedation at most and then usually using only intravenous agents. However, beyond the dental chair, the use of inhalational methods for relative analgesia and sedation can still result in significant exposure. In these techniques, high flow rates of nitrous oxide can be used (occasionally together with low concentrations of volatile agents).



Figure 18.2 Nitrous oxide analyzers. On the right, a Medigas PM 3010 portable analyzer and on the left, a Dräger sampling tube and badge.

MEASUREMENT OF POLLUTION

The extent of pollution in the theatre environment is now quantifiable. It may be measured by various methods, some of which are described below.

Operating theatres

With the introduction of low-cost non-dispersive, portable infrared analyzers, trace quantities of anaesthetic agents can be measured continuously. A direct reading analyzer (Fig. 18.2) enables spot measurements to be taken at different sites, allowing the background level of nitrous oxide in a room to be assessed. Instant results of nitrous oxide levels (in the range of 0–1000 ppm with a resolution of 5 ppm) are displayed in real-time or as an 8 h TWA. An alarm protects personnel against excessive levels of exposure.

Theatre personnel

Individuals can be issued with sampling tubes for nitrous oxide (Fig. 18.2) and sampling badges for volatile anaesthetics (Fig. 18.3) that are worn for approximately 8 h.



Figure 18.3 Dräger sampling badge for analyzing volatile agents.

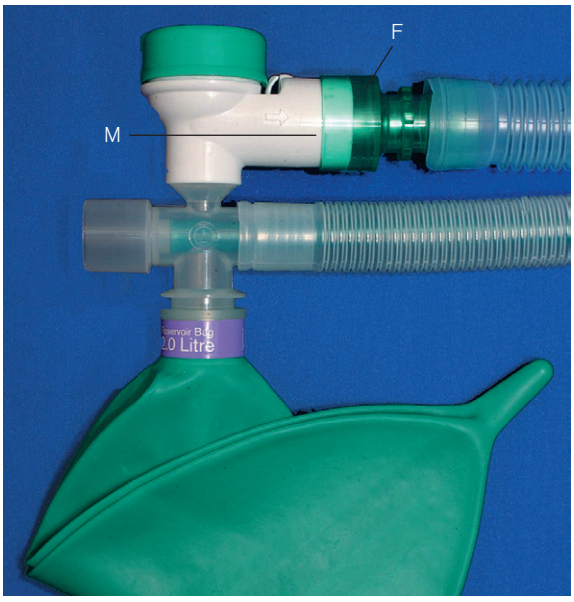


Figure 18.4 A collection system showing a shrouded APL valve encased in a gas scavenging collector system and terminating in a 30 mm conical male taper (M), and a 30 mm female conical taper (F) and 30 mm corrugated tubing for linking the former to a transfer system.

They are placed at shoulder height and the pollutants are adsorbed onto the material in the sampler in proportion to their concentration in the ambient atmosphere. For volatile agents, the material is based on activated charcoal, whereas for nitrous oxide, a molecular sieve is used. At the end of the passive sampling period, the samplers are sent to a specialist laboratory where the pollutants are measured using a gas chromatograph linked to an infrared detector.

Biological monitoring of post-volatile anaesthetic exposure, using urine samples analyzed by gas chromatography-mass spectrometry coupled with static headspace sampling, has been shown to be another useful tool to monitor the extent of exposure.[18]

SCAVENGING SYSTEMS

A scavenging system transports waste gasses and vapours from a ventilator or breathing system and discharges them at a safer remote location. It includes several components, namely:

- *a collecting system*, which conveys waste gasses from the breathing system to a transfer system
- *a transfer system*, which consists of a section of flexible wide-bore hose linking the collecting system to the receiving system
- *a receiving system*, which behaves as a reservoir to store surges in the flow of waste gas. From here, these gasses have to pass via disposal tubing to a disposal system
- *a disposal system*. This then transports the waste gasses to a site on the outside of a building away from populated areas.

Two or more of these items may be embodied in a single item of equipment.

Waste gas normally passes through the collecting and receiving system to the disposal system, using only the power generated in exhalation by the elastic recoil of a patient's lungs. At this stage there is little difference between the various systems employed. It may then pass through the disposal system using this same power (*passive scavenging*). However, it may be assisted by some form of gas or electrically powered apparatus, which generates a sub-atmospheric pressure (*active scavenging*). Only systems that employ active scavenging are able to deal with the wide range of expiratory flow rates (30–120 L min^{-1}) seen in anaesthetic practice, especially when certain ventilator systems are used. Active systems are, therefore, the only ones that can be recommended – provided that they also meet certain specification and performance criteria (BS EN 740:1999).[19]

The collecting system

This has two components:

1. A 30 mm male conical connector (labelled M in Fig. 18.4) that is fitted either to the expiratory port of a ventilator, the demand valve (for Entonox)

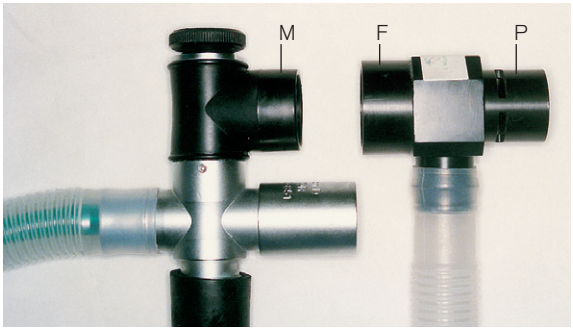


Figure 18.5 A collection system (M) and transfer system (F) incorporating an overpressure relief valve (P) (set at 1 kPa) in the female component.

or to the APL valve of a breathing system. The version that fits the APL valve shrouds all the exit apertures on the body of the valve enclosing them in a gas-tight fit.

2. A 30 mm female conical connector (labelled F in Fig. 18.4) that fits over the male connector to form a gas-tight fit and is attached to the patient end of the transfer system.

Having two components in the collecting system allows the female part to be detached and reattached to different breathing systems as required. The selection of a unique 30 mm taper for this connection is intended to prevent other breathing system components from being attached to it in error.

The collecting system may also house an overpressure relief valve, which is normally set to blow off at 1 kPa (10 cm H_2O). This device (Fig. 18.5) prevents excessive pressure building up in the breathing system if the scavenging system becomes obstructed; for instance due to crushing or kinking of the transfer tubing.

The transfer system

This consists of a length of wide-bore, kink-resistant tubing that joins the collecting system to the receiving system.

The exhaled gasses emerge intermittently from the breathing system, their volume and flow pattern varying according to the type of apparatus in use. For example, with spontaneous respiration there may be a fresh gas flow of approximately 8 L min^{-1}, which must be scavenged. However, the peak flow rate during the period at which the APL valve is open may be much higher (up to 45 L min^{-1}). Furthermore, some ventilators that have gas-driven bellows discharge both driving gas and exhaled gas into the scavenging system. This might well produce intermittent flow rates in excess of 100 L min^{-1}. It is important, therefore, to match the performance of the rest of the system to that of the anaesthetic equipment.

However, it would be dangerous to attach the collecting apparatus directly to a scavenging system that was actively extracting gas in excess of 120 L min^{-1}, as this would suck gas out of both breathing system and patient.

The solution is to install a safety device (*receiving system*).

The receiving system

The receiving system (Fig. 18.6) consists of:

- *a reservoir* (normally a rigid material cylinder) for the expired and driving gasses, from which they are passed to the disposal system. This may also temporarily store this gas if the extraction rate falls below the necessary level
- *an air break*. The reservoir is open-ended at its base to allow entrainment of air when there is insufficient expired gas. This prevents the transmission of sub-atmospheric pressure from the scavenging system to the patient. It also provides an emergency escape route for the gas should the scavenging system fail
- *a flow indicator* to show that the unit is working when connected to an active disposal system. There is normally a clear Perspex window sited near the top of the reservoir in which a coloured float appears when the extraction rate is normal (i.e. 120 L min^{-1}). When the flow drops below 80 L min^{-1} this disappears from view
- *a filter* sited in the base of the unit to prevent debris entering and blocking the system
- an entry port on the side and an exit port on the top of the container.

The receiving system is connected to the disposal system via a wide-bore hose that is sufficiently strong to prevent collapse from the sub-atmospheric pressure within it. The hose terminates in a probe, which houses a screw-fit connection to the disposal system socket (terminal unit). The latter has a valve that is normally closed, but opens when the male probe from the receiving unit is connected to it and screwed in (Fig. 18.7). The terminal unit may be sited on a wall or pendant.

The disposal system

Active disposal systems

The sub-atmospheric pressure required to power the disposal system is usually provided by an exhauster unit (Fig. 18.8). This works in a similar fashion to a fan and requires a low level of maintenance and no lubrication. The size of the unit depends on the number of scavenging sites to be

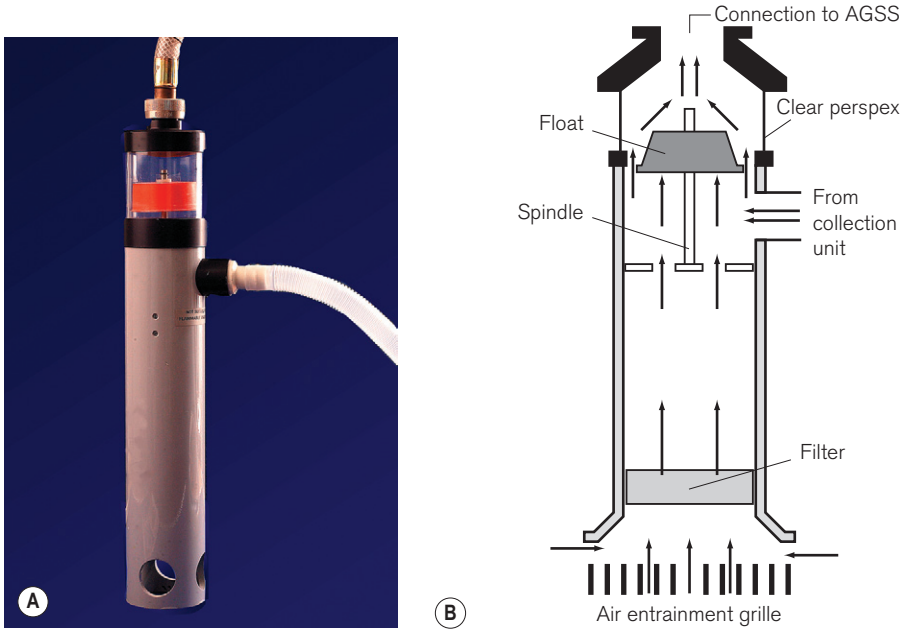


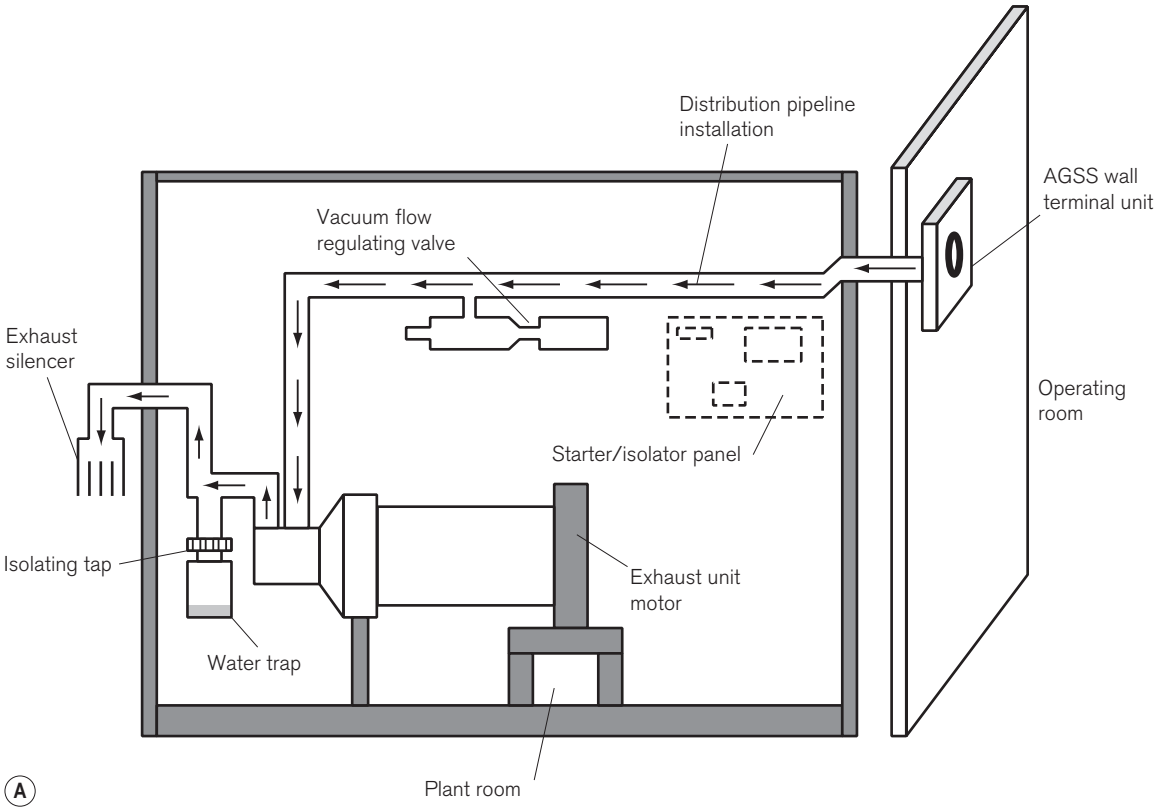
Figure 18.6 **A.** An Ohmeda receiving system for use with active gas scavenging system (AGSS). **B.** Internal arrangement of the system.



Figure 18.7 The probe from a receiver unit can be inserted into either: **A.** a wall-mounted terminal or **B.** a ceiling-mounted terminal.

supplied. Large exhauster units can provide waste gas flow rates of up to 2400 L min^{-1}, servicing 20 sites. Large sites often have a 'duty' and a 'standby' unit, which are linked. The standby unit operates automatically if the duty unit fails, as well as during periods of high demand. Although the exhauster unit is sited outside the operating theatre suite (sometimes a considerable distance away), the operating control switch is sometimes located within the theatre suite.

Pressure fluctuations within the disposal system are controlled within precise limits by a vacuum/flow-regulating valve. It consists of an adjustable spring-loaded plate



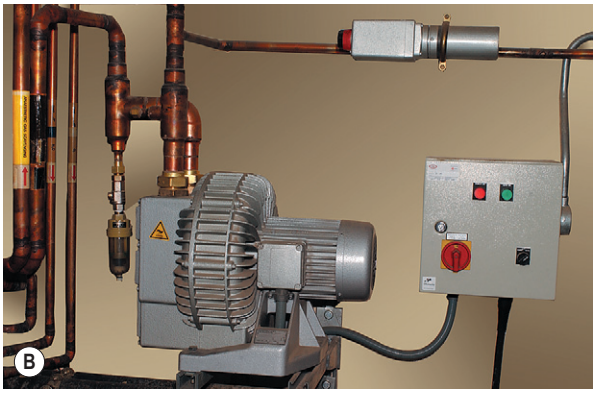




Figure 18.8 Active gas scavenging. **A.** Schematic diagram of plant room. **B.** The disposal unit in a plant room. **C.** Exhaust silencer.

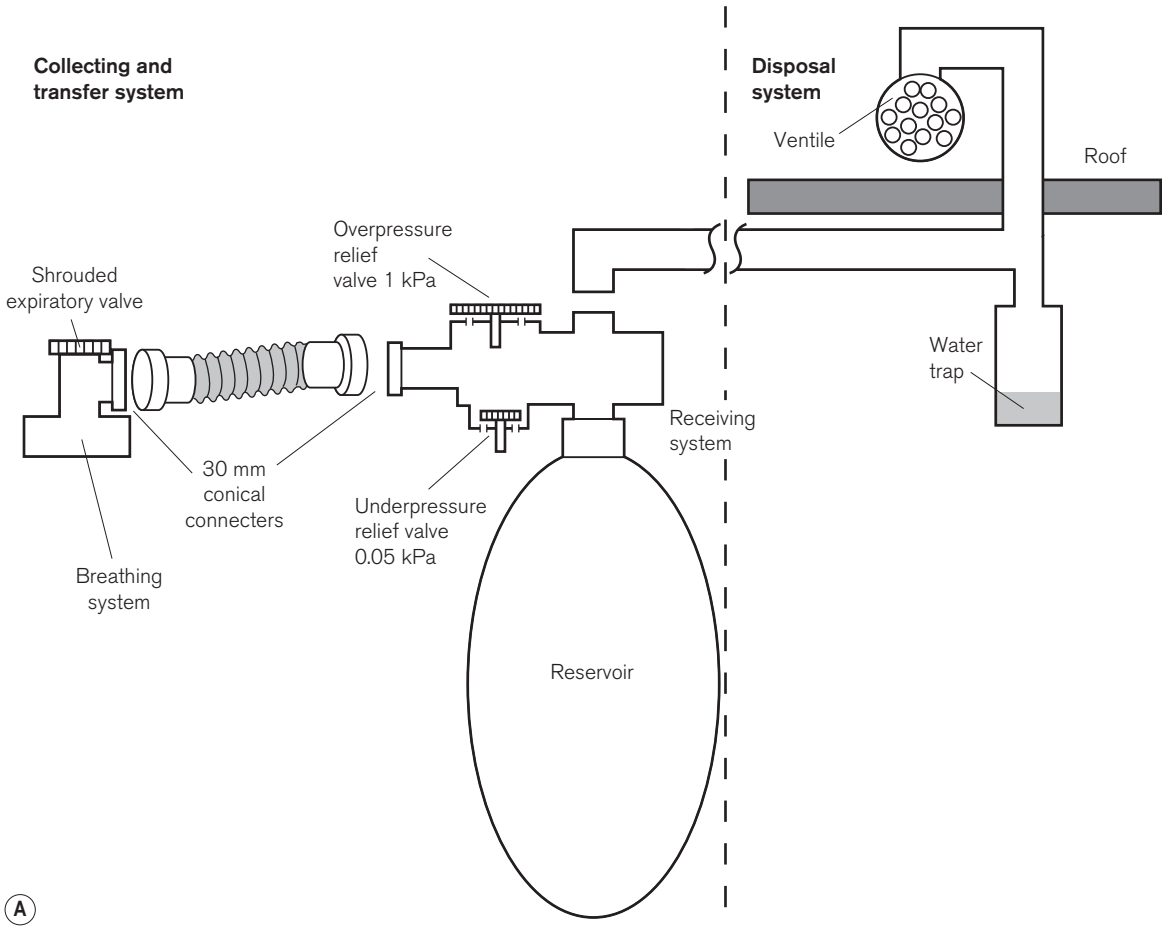


Figure 18.9 **A.** A passive scavenging system. **B.** A ventile for waste gas disposal emerging from the wall of an operating theatre suite.





Figure 18.10 **A.** Gas disposal via wall outlet on floor of theatre. **B.** Exhaust pipe communicating to the outside.

then attached to a special active gas scavenging system) can be sited close to a patient's face to remove pollutants. The funnel is supported by a series of levers. For efficacy, the device requires a calm patient lying under the optimal extraction zone of the device. In practice, an efficient non-recirculating air-conditioning system in these areas would be more appropriate.



Figure 18.11 The Cardiff Aldasorber.
Photograph courtesy of Shirley Aldred & Co. Ltd. Sheffield, UK.



Figure 18.12 A collection system for use in recovery rooms.

REFERENCES

1. Brown AC, Canosa-Mas CE, Parr AD, Pierce JM, Wayne RP. Tropospheric lifetimes of halogenated anaesthetics. *Nature* 1989;**341**:635–7.
2. Halsey MJ. *Occuaptional exposure to anaesthetics. Anaesthesia Rounds*. Abingdon: The Medicine Group (Education) Ltd; 1996.
3. Albritton DL, Auchamp PJ, Mergie G, Watson T. *Scientific assessment of ozone depletion: 1998, World Meteorological Organization Global Ozone Research and Monitoring Project Report No. 44*. Geneva: WMO; 1999.
4. Langbein T, Sonntag H, Trapp D, Hoffmann A, Malms W, Röth EP, et al. Volatile anaesthetics and the atmosphere: atmospheric lifetimes and atmospheric effects of

halothane, enflurane, isoflurane, desflurane and sevoflurane. *Br J Anaesth* 1999;**82**:66–73.

5. Ravishankara AR, Daniel JS, Portmann RW. Nitrous oxide (N_2O): the dominant ozone-depleting substance emitted in the 21st century. *Science* 2009;**326**:123–5.
6. Vaisman AI. Working conditions in surgery and their effect on health of anesthesiologists. *Eksperimental'naia Khirurgiia i Anesteziologiia* 1967;**3**: 325–30.
7. Bruce DL, Bach MI. Effects of trace anaesthetic gasses on behavioural performance of volunteers. *Br J Anaesth* 1976;**48**:871–6.
8. Guirguis SS, Pelmear PL, Roy ML, Wong L. Health effects associated with exposure to anaesthetic gasses in Ontario hospital personnel. *Br J Ind Med* 1990;**47**:490–7.
9. Vouriot A, Gauchard GC, Chau N, Nadif R, Mur JM, Perrin PP. Chronic exposure to anesthetic gasses affects balance control in operating room personnel. *Neurotoxicology* 2005;**26**:193–8.
10. Rowland AS, Baird DD, Weinberg CR, Shore DL, Shy CM, Wilcox AJ. Reduced fertility amongst women employed as dental assistants exposed to high levels of nitrous oxide. *NEJM* 1992;**327**:993–7.
11. Deacon R, Perry J, Lumb M, Chanarin I, Minty B, Halsey MJ, et al. Selective inactivation of vitamin B12 in rats by nitrous oxide. *Lancet* 1978;**2**:1023–4.
12. Health and Safety Commission. *Anaesthetic agents: controlling exposure under COSHH*. Bristol: Health Services Advisory Committee; 1996.
13. Health and Safety Executive. *Control of substances hazardous to health. 5th ed. The Control of Substances Hazardous to Health Regulations 2002 (as amended) Approved Code of Practice and Guidance*. London: HSE Books; 2005.
14. Health and Safety Executive. *Occupational Exposure Limits.* Guidance Note EH40/96, 1996.
15. Health and Safety Executive. *Workplace exposure limits*. EH4O/2005, 2005.
16. Davenport HT, Halsey MJ, Wardley-Smith B. Bateman PE Occupational exposure to anaesthetics in 20 hospitals. *Anaesthesia* 1980;**35**:354–9.
17. National Institute for Occupational Safety and Health. *Criteria for a Recommended Standard: Occupational Exposure to Waste Gasses and Vapours, DHEW Publication No. (NIOSH) 77–140 Cincinnati*. Ohio, USA: NIOSH; 1977.
18. Accorsi A, Barbieri A, Raffi GB, Violante FS. Biomonitoring of exposure to nitrous oxide, sevoflurane, isoflurane and halothane by automated GC/MS headspace urinalysis. *Int Arch Occup Environ Health* 2001;**74**: 541–8.
19. British Standards. Anaesthetic workstations and their modules. Particular requirements. BS EN 740:1999. London: British Standards Institution; 1999.

FURTHER READING

Medical gas pipeline systems – Part 2: Anaesthetic gas scavenging disposal systems. BS EN ISO 7396-2:2007. Geneva: International Standards Organization; 2007.

Chapter 19

Infusion equipment and intravenous anaesthesia

Ali Diba

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Microprocessor-controlled infusion devices are now so ubiquitous that outside the operating theatre it is rare to see intravenous infusions administered without these. Such devices can be commanded in any number of units from ml h^{-1} to mass units of drug per unit patient weight per unit time. Within operating theatres the ease of use of a target-controlled infusion (TCI) as a mode of drug delivery has dramatically increased the use of intravenously maintained anaesthesia. From the viewpoint of the infusion device, TCI is, of course, simply another layer of calculations for its microprocessor. 'Open Label TCI' (where the device does not require the manufacturer's specially presented drug product to allow administration by TCI), has become a reality; and for more than one drug. TCI appears no longer to be seen by regulatory authorities as a peculiar licensing issue linking infusion device and drug manufacturer.

Increasing functionality obviously carries increased risk of drug maladministration and in mitigation the use of 'drug libraries' loaded as additional software onto infusion devices has become commonplace. These together with various other 'error traps' aim to reduce the opportunities for common errors.

Software revisions, access codes, data logging, alarms and the additional desire by manufacturers for miniaturization and for producing singular machines that can be made to behave effectively as different devices (e.g. TCI pump, critical care unit infusion pump, patient-controlled analgesia (PCA) device), render these machines complex, occasionally prone to software failure and with often cumbersome user interfaces. In this light, the popularity of simple elastomeric infusion devices is not difficult to comprehend.

EVOLUTION OF INFUSION SYSTEMS

It was once common to see doctors and nurses with watch in hand, converting infusion rates from 'duration of infusion' to drops per minute and adjusting the roller clamp on an intravenous 'drip' set. Infusates would run through too fast or too slow, because the plastic tubing altered shape or the downstream resistance in the intravenous cannula altered for a number of reasons. Often it was simply the calculations that were wrong.

This was superseded first by electronic *drip counters*, which were subsequently made to control the adjustable clamp on a gravity-fed giving set (*infusion controllers*) and, ultimately, by microprocessor-controlled *infusion pumps* (Fig. 19.1), able to generate flow irrespective of the effects of gravity, and incorporating many features such as sensing infusion line pressure and the facility for being programmed in a variety of units and even languages to give stepped infusions based on patient weight. The first devices to not rely on a drip counter used a syringe type cassette to reliably control the volumes delivered and were hence known as '*volumetric pumps*' to differentiate them (Fig. 19.2). This design is less common now, and most devices rely on a peristaltic mechanism to push fluid

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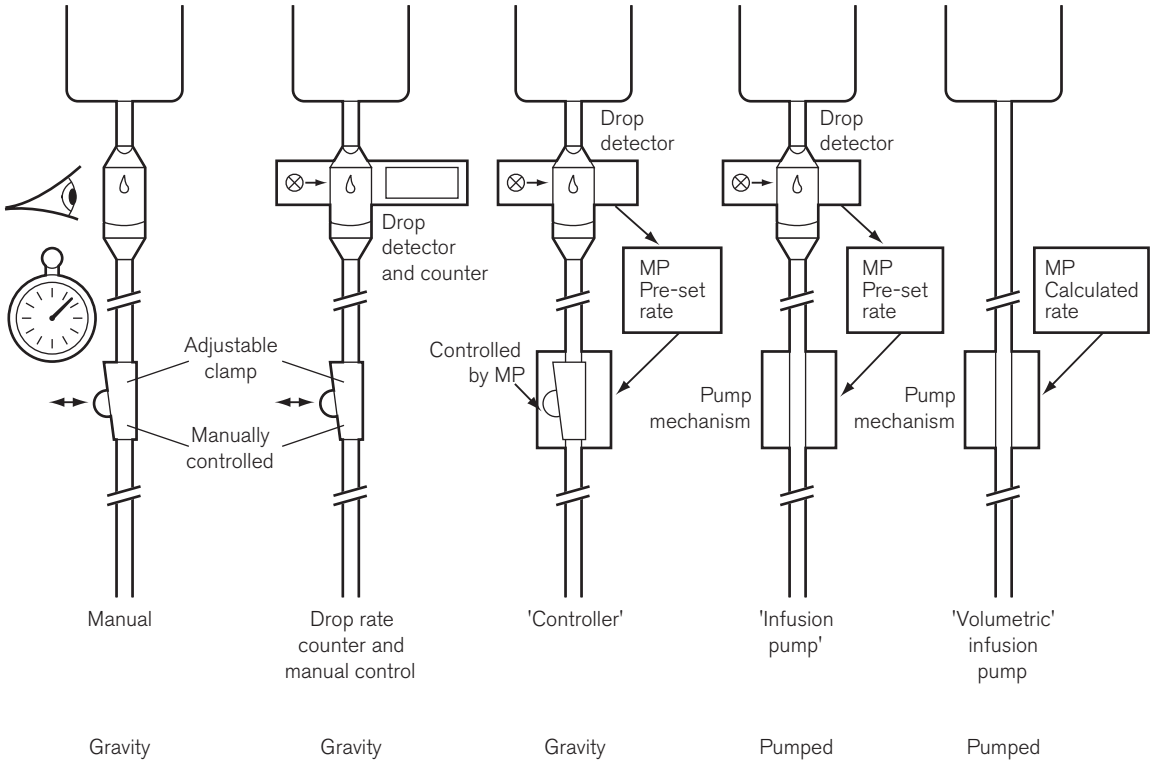


Figure 19.1 Evolution of infusion pumps, (MP = microprocessor).

along. The term should now probably be made obsolete to be replaced by 'infusion pump'.

Syringe drivers (also called *syringe pumps)* have traditionally been used for more accurate control of smaller volumes of infusion. The very first were simple clockwork devices designed to drive the plunger of a syringe at an even rate usually over 24 h. Syringe pumps now use the same technology as other infusion pumps and all these devices are considered together here.

1996 saw further progress with the commercial introduction of the Diprifusor system. This, a proprietary 'chip' added into a microprocessor-controlled syringe driver, allows the delivery of specially packaged Diprivan (propofol) by *target-controlled infusion (TCI)*, and is discussed in greater detail below and also again later in the chapter. Licensing issues initially hampered the arrival of '*open label TCI*' (see below), but at least four manufacturers now produce devices with open label pharmacokinetic algorithms capable of infusing at least the drugs propofol and remifentanil by TCI. Although it would be logical to use such systems for any drug (anaesthetic or otherwise) with a short half-life that needs to be given by continuous infusion, the development costs and licensing concerns for this route of administration for any given drug mean that it is only rarely commercially viable.

Microprocessor controlled/ software driven

Modern electrically powered infusion devices use a stepper motor (see below). This is controlled by a microprocessor which ultimately simply varies the flow rate of the device from between 0 to 999 ml/h (infusion pump) or 1200 ml/h (for a syringe driver, depending on syringe size) at any given time.

Depending on the intended purpose of the infusion device the processor can be made to accept commands from the user for controlling the infusion in many different ways:

- simple ml/h request giving rise to the simplest infusion pump, say for ward use
- unit of drug per unit patient weight per unit time with variable units used for each; for example mg/kg/h or µg/kg/min. Clearly when dosing drug in this manner, the drug concentration in use and the patient weight must be input to the pump to allow automatic calculation of the flow rate. Such pumps may be used in the intensive care unit setting, but are more often termed '*anaesthesia pumps*' as liberal discretion for choice of units is now usually only entrusted to anaesthetists

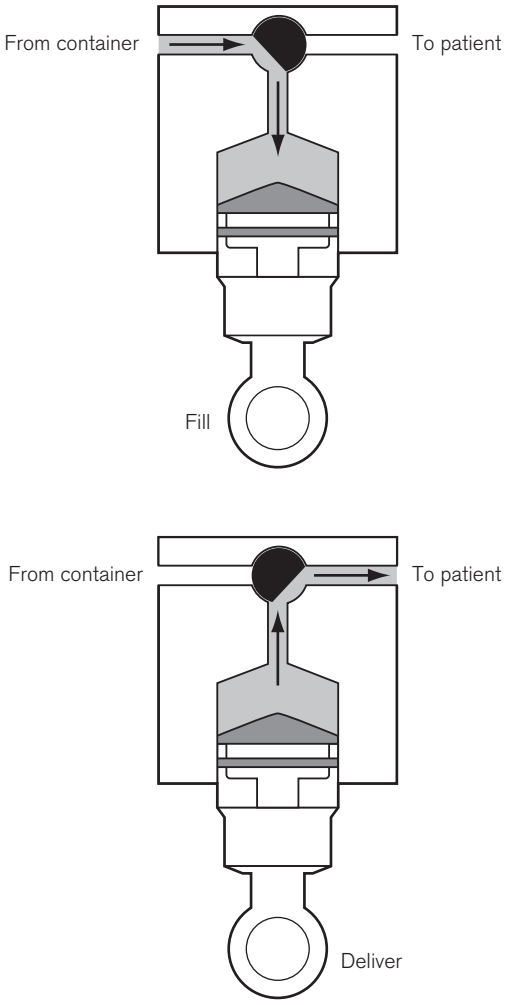


Figure 19.2 Principle of a cassette type ('volumetric') infusion pump.

- simple infusion rate with an additional bolus volume when commanded by a separately attached control handle, thus constituting a PCA pump
- the in-built processor (or an additional piggybacked processor) can contain algorithms for the pharmacokinetics of particular drugs, allowing automated drug delivery based on achieving any desired theoretical patient plasma or effect site concentration of that drug. Such a configuration is termed a 'target-controlled infusion (TCI) pump'. These systems are currently only commercially available for use in anaesthesia and are principally limited to four drugs: propofol, remifentanil, sufentanil and alfentanil. Patient variables such as age, sex, weight and height may need to be inputted to complete the algorithm. The algorithm, using this information together with the known drug concentration in the syringe, thereafter allows the anaesthetist to simply select a target patient concentration of drug. The pump automatically alters the infusion rate to most rapidly attain and maintain the calculated drug concentration in the patient at the set target. Such systems are very simple to use and obviate the need for complex calculations and detailed knowledge of pharmacokinetics by the user. They are largely responsible for a revolution in the administration of anaesthesia in a trend away from volatile agent maintenance towards total intravenous anaesthesia (TIVA).

Depending on the design of the chip these functionalities may be imbedded as hardware or firmware or added as software instructions. It can be seen how one basic infusion pump chassis can thus be made to give rise to effectively many different devices according to the microprocessor configuration. Each configuration or mode of use has its own attendant hazards in addition to those general to infusion pumps. Although all functionalities may be available in the one device, for risk management they may be selectively disabled through a restricted access menu such that only facilities that are needed in a clinical area are available to the user.

Modern devices are also able to communicate with centralized automated data archiving systems to automatically record the administration rate of a drug alongside the output of patient physiological monitors.

Simple infusion systems

In the operating theatre where there is closer observation of the patient's hydration and circulating volume as necessitated by their rapidly changing status, most intravenous fluids are still administered under gravity from flexible plastic containers using single-use fluid administration sets. These are of several types:

- simple fluid administration, no filter, droplet size approximately 15 drops/ml (0.07 ml)
- blood and fluid administration with clot filter at about 200 μm mesh size. These giving sets use a larger-bore tubing and a double drip chamber containing a float: by squeezing the bottom chamber, the float jams the inlet and acts as a one-way ball valve allowing fluid to be pumped. Droplet size is usually approximately 15 drops/ml
- burette, 100–150 ml in volume, droplet size either 15 drops/ml or, for paediatric usage, 60 drops/ml. A flap or ball valve at the bottom of the burette prevents air entering the drip chamber when the burette is empty
- platelet giving sets, designed to reduce the risk of aggregation in the giving set as would occur with conventional blood giving sets.

Some giving sets are now also designed so as to also be compatible with volumetric infusion pumps. This necessitates a narrower-bore tube made from softer plastic to function with the peristaltic pumps (see below). Flow rates are, therefore, lower and the tubing is less kink-resistant. These sets usually have a 15 µm filter at the base of the drip chamber and are not suitable for infusing blood or for use in adult resuscitation.

The rate of infusion in a simple gravity fed system depends on:

- the height of the fluid container above the infusion site
- the resistance to flow caused by the infusion set
- occlusion of the tubing from a rate controlling device
- the physical properties (i.e. viscosity) of the fluid to be administered
- the bore of the intravenous cannula
- the hydrostatic pressure in the veins of the patient.

The manufacturers of giving sets quote the size of drops as number of drops per millilitre, usually between 10 and 60, but it must be remembered that the actual volume of the drops depends on the physical properties of the fluid being administered.

Rapid infusion

When rapid infusion of blood or other fluids is required, the rate of administration may be increased by the use of an inflatable pressure bag (Fig. 19.3), which may even be contained in a rigid box to increase the speed at which pressure may be applied.

When fluids are administered at a rapid rate, provision should be made for warming them to body temperature, otherwise significant cooling of the body may ensue (see Chapter 30).

PRINCIPLES OF INFUSION DEVICES

Pumped infusion systems overcome the variation in infusion rates caused by changes in back-pressure, tubing resistance and the vertical height of the fluid container above the patient. Hence, they have to be powered by some form of motor, which must be coupled to a mechanism for driving the fluid

The stepper motor

The driving force in the majority of infusion pumps and electronic syringe drivers is provided by an electronic stepper motor, which is directly controlled from a digital microprocessor system. The speed of a conventional electric motor driven from either an AC supply or a DC supply may vary with mechanical load, the voltage or the frequency of the supply. It is, therefore, difficult, without electronic feedback, to control such a motor accurately. The stepper motor (Fig. 19.4) is designed so that a series of pulses applied to the stator windings of the motor cause the shaft to rotate by a fixed amount for each pulse, typically 1.8°, 2.5°, 3.75°or 7.5°, irrespective (within certain limits) of the load. Infusion systems are designed so that a pulse generator, whose output frequency is varied by the

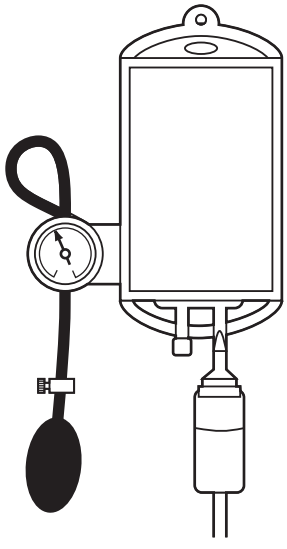


Figure 19.3 Pressure infusion device.

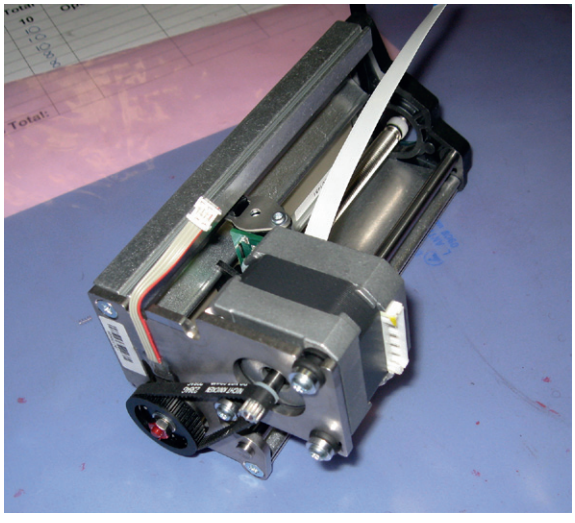


Figure 19.4 The stepper motor and lead screw assembly of an Alaris PK syringe pump during manufacture.

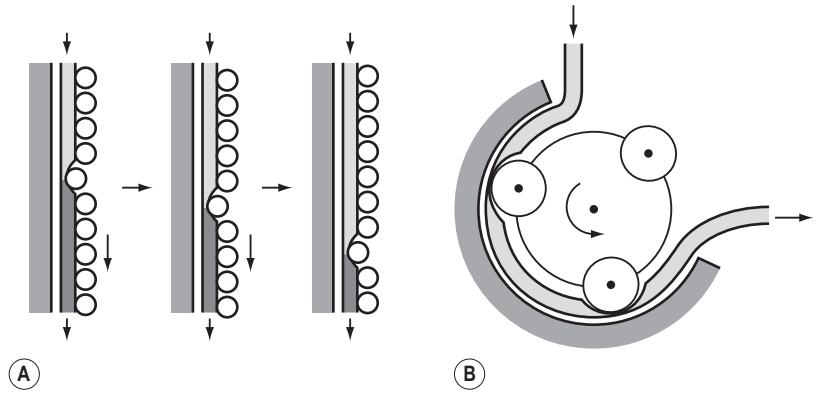


Figure 19.5 **A.** Linear, and **B.** rotary peristaltic mechanisms.

microprocessor, can produce accurate control of an infusion by varying the speed of the stepper motor.

Infusion pumps

Cassette type

Originally, the most accurate (and expensive) infusion pumps used syringe type cassettes (see Fig. 19.2) and were referred to as volumetric infusion pumps. This, like the newer peristaltic pumps, is driven by a stepper motor controlled directly by a microprocessor. The volume of the cassette is typically about 5 ml, with the dedicated disposable 'syringe cassette' for each manufacturer's pump being supplied separately as a sterile product. A valve operating in harmony with the piston directs flow from the infusate bag to the reservoir or from there to the patient. Fluid is drawn rapidly from the reservoir bag into the cassette in less than 1 s. The valve is then actuated such that on the piston upstroke the contents are expelled at the required rate into the patient, and the cycle is repeated. Although effectively this produces an intermittent flow, it also gives overall extremely accurate infusion rates, with only infrequent 1 s interruptions.

These are seen much less frequently now owing to the expense of the disposables and the feasibility of getting good accuracy using simple intravenous giving sets in modern peristaltic pumps.

Peristaltic pumps

The principle of the peristaltic infusion pump is shown in Fig. 19.5. The tubing of a giving set is compressed by a series of rotating rollers or by a wave of mechanical 'fingers' or cam followers. The section of tubing in the peristaltic mechanism must be hard wearing, of known and consistent internal volume, and have no memory after compression so that it easily fills on being released. Depending on the manufacturer, specific proprietary tubing with dedicated fittings may be needed to allow it to be loaded into the pumping mechanism. Precision silicone tubing is often used in this section of the giving set.

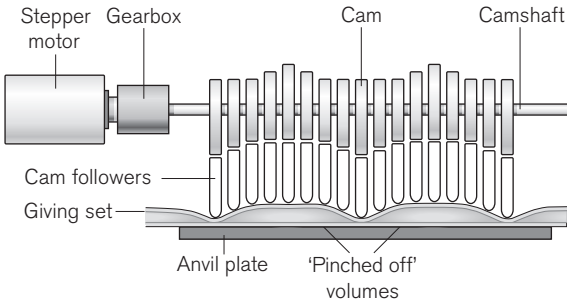


Figure 19.6 The rotary motion from the motor is translated into a linear peristalsis by the use of cams and cam followers.

Rotary peristaltic pumps are now more often seen in use for the less demanding requirements of enteral feeding rather than intravenous administration.

Linear peristaltic mechanisms allow much easier loading of the infusion tubing and are now by far the commonest design of infusion pump. The driving force is again a stepper motor. The rotary motion from the motor is translated into a linear peristalsis by the use of cams and cam followers as shown in Fig. 19.6.

Because such infusion pumps have the theoretical capacity to inject limitless quantities of air into a patient should air ingress occur upstream of the pump (for example due to an empty infusion bottle), these devices incorporate sophisticated ultrasonics (Fig. 19.7) or optics-based 'air in line' detection systems capable of sensing air bubbles as small as 0.1 ml volume. These are usually placed downstream of the pump mechanism. Further protection is conferred by setting target delivery volumes smaller than the volume in the bag of infusate.

To detect obstructed or extravasated catheters, electrically powered infusion devices must have some measure

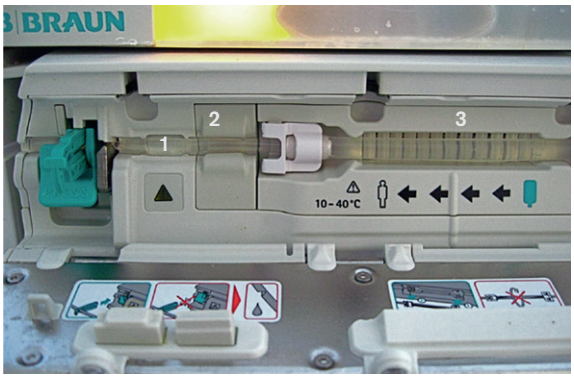


Figure 19.7 Braun Infusomat Space infusion pump with front cover opened to show (1) line pressure sensor, (2) ultrasonic air sensor, (3) peristaltic mechanism with precision silicone tubing loaded.



Figure 19.9 Graseby battery operated syringe drivers calibrated in mm per h and mm per 24 h.
Photo courtesy M Stewart, Graseby Medical Ltd.

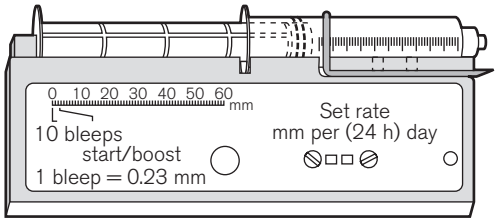


Figure 19.8 Graseby syringe driver.

of the pressure generated in the infusion line beyond the device. Peristaltic intravenous pumps, therefore, use a sensing piston pressing on the infusion line immediately downstream of the pumping chamber. This is calibrated to indirectly measure line pressure and can be programmed to alarm for occlusion at different pre-set levels.

Syringe drivers

There continues to be a range of small simple battery-operated syringe drivers (Fig. 19.8). The driving mechanism is a miniature DC motor that is switched on and off intermittently and drives a screw-threaded rod (lead screw), which is linked to the syringe plunger, causing its advancement. They may have a variable rate that is altered by adjusting a recessed control using a small screwdriver. These pumps are small and light enough to be worn in a holster by an ambulant patient and are now used chiefly for narcotic infusions for the relief of cancer pain. Great care must be taken in calculating drug dilutions and to ascertain that the correct units are used for setting the infusion rates, as the pumps are available in different models with rates set either as mm per 24 h or mm per h of plunger movement (Fig. 19.9).

Virtually all other syringe drivers for hospital use and particularly those used in intensive care and anaesthesia use microprocessor controlled stepper motors, again connected to the syringe plunger by a carriage on a lead screw (Fig. 19.10). Thus, each pulse applied to the stepper motor causes the advancement of the syringe plunger by a known amount. The pulse generator may be calibrated from 0.1 ml h^{-1} to 1200 or occasionally 1800 ml h^{-1}, the higher rates being used only for delivering a bolus (often of predetermined volume) or for purging the infusion line.

Syringe pumps (the term is synonymous with syringe drivers) are now designed to automatically recognize a variety of syringes by virtue of the calibre of the barrel using some form of spring-loaded arm; some manufacturers' models nonetheless require manual confirmation of the detector. Infusion line pressure (and empty syringe detection) is calculated indirectly from the force acting on the syringe plunger by sensors, which may be incorporated into the carriage or lead screw assembly (Fig. 19.11). This is a more popular option than the use of specialized infusion sets with in-built diaphragm and corresponding transducer housing on the syringe pump which remains largely confined to pumps used on neonatal ICUs. The facility in some devices to also alarm for *low* infusion line pressures is intended to allow recognition of disconnection of an infusion line (with the aim of, for example, preventing awareness in intravenous anaesthesia).

Rechargeable batteries

Although mains-driven, electrical infusion devices must have battery back-up both to cover mains failure and for patient transfer and emergency situations. The performance of the in-built rechargeable batteries is an important consideration when purchasing such equipment, but it must be remembered that this is also influenced by the battery maintenance procedures. Poor battery life can

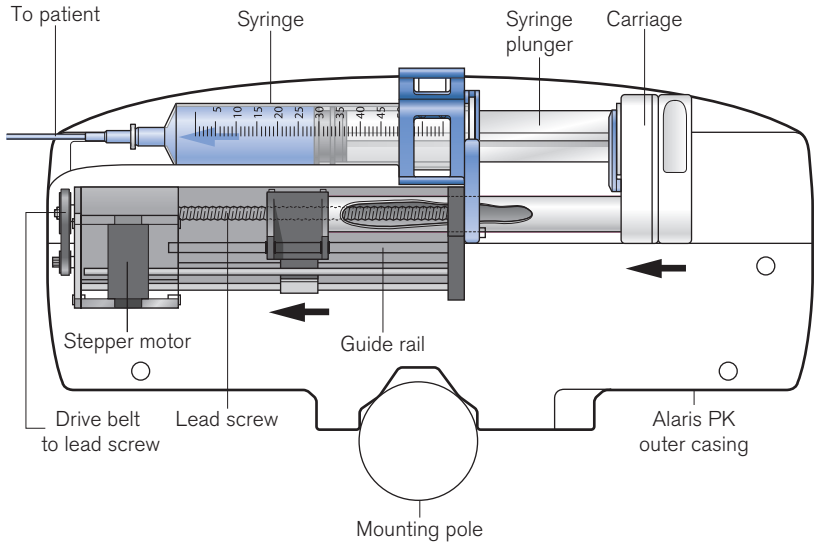


Figure 19.10 Syringe driver with microprocessor-controlled stepper motor, connected to the syringe plunger by a carriage on a lead screw.



Figure 19.11 A piezoresistor is bonded onto the metal chassis which carries the lead screw of an Alaris Asena syringe pump. The distortion (invisible to the eye) of the steel plate caused by the force acting on the lead screw causes a change in resistance which can be calibrated to be read as a pressure within the syringe and giving set.

render otherwise excellent devices unreliable and unusable. Pumps should be kept connected to the mains when not in use and batteries should be replaced appropriately. Microprocessor-driven infusion devices are susceptible to bizarre error conditions when rechargeable batteries begin to fail.

In common with many rechargeable batteries, *nickel-cadmium* (NiCd) rechargeable batteries should be periodically run down completely to prevent the development of 'memory', which renders them unable to discharge their full capacity. NiCd batteries are gradually being replaced by *nickel-metal hydride* (NiMH) batteries for environmental reasons (cadmium is a toxic heavy metal). In comparison to NiCd batteries, NiMH batteries have a higher-energy density, i.e. they can hold more charge per unit weight, but have a more limited service life. They are similarly prone to memory problems and need appropriate maintenance. *Lead acid* batteries, also called '*sealed lead acid*' to designate portability and to differentiate from the *flooded* type used in cars, have no 'memory', are cheap and reliable, but have long charging times. They are most often found on portable equipment such as ventilators and other heavy devices (e.g. wheelchairs and 'uninterruptible power supply' systems). Lead acid batteries, conversely, suffer by being allowed to fully discharge and must not be stored in this state as the process of *sulphation* can render them unusable. Lithium-ion (and lithium polymer) batteries have a high-energy density and no memory but are very expensive. The technology is currently confined largely to portable personal electronic equipment such as mobile telephones.

Battery maintenance is difficult in the hospital setting. Medical device batteries obviously cannot be safely run down whilst in use. Similarly, encouraging even partial discharge of a battery decreases the safety margins in the event of power failure or the need for transportation of a device.

Safety

Microprocessor-driven infusion devices are used for the administration of many potent drugs with narrow

therapeutic windows where maladministration can have lethal consequences. The drugs are used in a variety of dilutions and dosed in units that can vary by several orders of magnitude (ηg/ml, μg/ml, μg/kg/min, mg/kg/h). The devices themselves are highly versatile and capable of being instructed to perform in many different manners. Given these factors it is evident that there is significant potential for user error with disastrous consequences.

The devices themselves may also malfunction, albeit infrequently. The pump processor monitors many aspects of the device's performance in order to detect malfunction and to make operation safe. Though rare, glitches in the software may under certain circumstances cause over- or under-infusion. Software issues are more commonly seen as a stopped infusion when the processor receives apparently conflicting messages from different sources in the device, which then is made to *fail safe* with appropriate alarms and error codes. An unwanted termination of infusion can be equally dangerous – if, for example, the patient's circulation is dependent on vasoactive drugs.

The machines have built-in alarms for occlusion, low battery, mains failure, disengagement of drive mechanism, failure to load infusion set and other common fault conditions. In spite of this, they remain high-risk devices capable ultimately of delivering drugs dangerously: they have a recognized associated morbidity and mortality. In at least 27% of the 1495 incidents involving infusion pumps reported to the Medical Devices Agency in the UK between 1990 and 2000, the cause was found to be user error (including failure to maintain the device appropriately).[1] Only in 20%, were problems device-related with issues such as performance, degradation, quality assurance and design and labelling. In the 53% of cases where no cause was established it is likely that a very large number represent user error. The MHRA report for the year 2009 again reports that infusion and feeding pumps 'continues to be one of our busiest areas' with 375 adverse incident reports.[2] The user must, therefore, be ever vigilant and particularly aware of the following problems:

- ***Unitary programming errors.*** The simplest and most common error is a mistake or slip in selecting the correct dosing unit or drug concentration, e.g. mg/kg/min instead of μg/kg/min or μg/ml instead of mg/ml. These issues are addressed to a large extent by the use of drug libraries (*v.i.*).
- ***Wrong drug errors.*** Where more than one infusion pump is used particularly when they are controlled through a common interface it is relatively easy to confuse the drugs. Propofol available as identical looking 2% and 1% solutions is also easy to confuse especially as one manufacturer of open label TCI pumps allows a change in concentration during a TCI (Fresenius) and one does not (Alaris), and hence does not prompt for concentration when new syringes are loaded. The Diprifusor system automatically senses the drug in the prefilled syringe and hence does not have this problem but this may now predispose doctors to errors when using other TCI systems.
- ***Failure to restart infusion.*** This is a very common error after refilling syringes during intravenous anaesthesia. Software changes across generations of the same device may require additional key presses for the same function (Diprifusor) and can contribute further to this error.
- ***Siphoning.*** This is the term used to describe the uncontrolled flow of fluid from a syringe into the patient under gravity. Modern drivers clamp the plunger and have a detector or mechanism built in to protect against incorrectly mounting the syringe plunger. It is best, even with modern designs, to not have the syringe driver higher than the patient, as small amounts of siphoning can still occur. Anti-siphon valves – essentially a one-way valve with a high opening pressure – may be incorporated into syringe pump giving sets (see below, Infusion lines). The possibility of inadvertent administration with infusion pumps as a result of administration sets not being properly clamped off when the pump is not in operation or when the infusion set has not loaded into the device properly still exists but should be largely designed out in new devices.
- ***Reflux.*** Where multiple infusion lines are connected to a single intravascular device and there is a distal obstruction it is possible for drugs from pumped lines to reflux up an attached gravity fed line. It is imperative that such lines have anti-reflux valves incorporated (see below, Infusion lines). It is worth stating that this fault situation can and does also frequently occur when drugs are injected into side ports of intravascular catheters or simple giving sets administering fluids by gravity feed. (Classically muscle relaxants injected into giving sets with blocked intravascular catheters, which are then cleared by postoperative recovery staff, resulting in a paralyzed awake patient.)
- ***Common deadspace and multiple infusions.*** See 'Infusion lines' below.
- ***Infusion of air.*** More commonly now due to unprimed infusion lines.
- ***Extravasation injuries.*** Continued infusion after a cannula has 'tissued' or become extravascular are not prevented by the use of low infusion line pressure limits.
- ***Electrical hazards.*** These are present as with all powered devices, but the common liquid spills on to infusion pumps can be particularly dangerous in the presence of cracked machine casings or frayed mains electricity leads.
- ***Software revisions in microprocessor-driven devices.*** These can produce a wholly new machine within the

familiar appearance of the old. Although features may be added or improved, there is also the possibility of introducing new problems and errors, particularly for those familiar with previous versions. Manufacturers should treat all but the most trivial software revisions as new devices and issue new instruction and training/maintenance manuals.

Because of the huge flexibility of microprocessor-controlled infusion devices, it is increasingly important for users of these devices to have familiarized themselves specifically with the features and functions of each model before clinical application. Programming errors are very common and may potentially result in lethal overdosage. These devices are not always entirely intuitive to use, errors are commonplace with, for example, the 'hands free bolus' facility (a pre-programmable bolus dose that does not require the button to be kept depressed during delivery), which may give the option of a variety of unexpected units and infusion rates.

Error traps and drug libraries

Error traps refer to systems set up within the microprocessing logic of the pumps to prevent certain errors. For example, pumps left switched on with infusion rates or doses programmed, but where the infusion is not started after loading an infusion set or syringe, may emit a low priority alarm to draw attention to this. An audible signal spanning the delivery of a 'hands free' bolus is common and particularly useful in anaesthesia, allowing recognition of a misprogrammed dose that does not take the expected time to deliver. Such features should not be disabled. In TCI mode where the specific drug algorithm has to be chosen, it is standard to have a preset range and limit for maximal target concentration to prevent accidental overdose. These limits may be as both a *soft limit* (where a confirmatory key press after a warning message allows the higher values) and a *hard limit* where higher values cannot be chosen. Clearly these limits are specific to the drug in use. For this reason it has become standard to have drug libraries programmed into pumps. Here, for any infusion, not just TCI or PCA, the user is obliged to choose the drug from a database which then gives default units and ranges for dilution and dosage, usually in the form of both hard and soft limits.

Drug libraries (Fig. 19.12)

- These are pre-programmed usually through a personal or networked computer.
- Setting up is a laborious process often requiring an upper and lower limit for each drug with regard to: patient weight, drug concentration, bolus size, bolus rate and infusion rate.
- They must be set up with great thought and attention to detail so as to:
  - be comprehensive, i.e. allow full range of drug and dosage requirements that may be needed
  - militate against inadvertent maladministration by having tight dosage limits and a restricted number of drugs to prevent confusion
  - not cause systematic and institution wide *forced errors*, e.g. wrong dosing unit or range pre-programmed.
- They need careful delineation of responsibility for maintenance, e.g. named *super-users* or the manufacturer, so that it is clear when and by whom changes are made and who carries ultimate responsibility.
- They are particularly appropriate where the user of the device is not the drug prescriber as in ward areas and the ICU.
- They do inherently decrease flexibility of infusion devices. Users in anaesthetic areas often programme in a series of 'unnamed' drugs administered in a number of differing mass units per unit patient weight per units of time to allow for unexpected eventualities. Pumps working through libraries are intrinsically slower to set up for use, especially as they also often do not have a numeric keyboard to allow direct entry of values. This may be a significant issue in anaesthetic areas with a high turnover of cases.

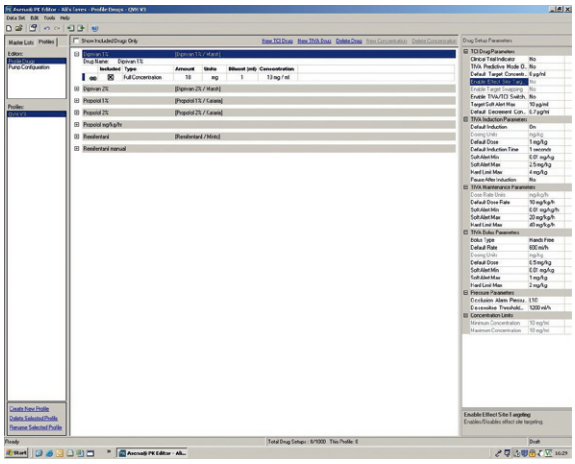


Figure 19.12 Sample page from a drug library set-up programme, the Alaris Asena PK editor.

Bar code scanners

Where drugs can be purchased or prepared in advance with imprinted bar codes or machine readable code, it is possible to combine this with the in-built library of the infusion device to give a simpler to operate and more error-proof system for drug administration (Fig. 19.13). In conjunction with connection to automated record keeping systems such a set-up also allows high-quality record keeping with obvious associations between



Figure 19.13 Alaris OSIS (optical syringe identification system) barcode scanner for automated drug recognition. *Image courtesy of CareFusion.*

medication changes and clinical effect. Such systems do currently exist but are not widespread in UK practice.

Line pressure limits

The line pressure limit at which devices alarm for occlusion can usually be altered; for some models this is a user function whilst on others it must be performed by biomedical engineering departments. Limits must be set such that nuisance alarms do not occur due to the resistance of the giving set or 'stiction' at the syringe barrel/plunger interface. High-pressure limits, however, have two major disadvantages:

1. It will take longer to alarm for occlusion (alarm time is increased) particularly at low infusion rates and at the start of an infusion – most commonly due to leaving a stopcock closed.
2. On release of the obstruction a proportionately larger bolus will be delivered to the patient; this may be clinically significant.

Electromagnetic interference

Microprocessors are susceptible to electromagnetic interference (EMI) from wireless telecommunication devices such as mobile telephones. EMI can cause serious malfunction in medical devices. The strength of the electromagnetic field is proportional to the power output of the telecommunication device and inversely proportional to the square of the distance from the source. The possibility of EMI is greatest with emergency radios (as used by ambulance, police and fire services) followed by security radios (as used by hospital maintenance and portering staff) and, finally, by mobile phones (both analogue and digital). The Medical Devices Agency of the UK reports 41% of the medical devices they tested suffered interference from emergency radio handsets at a distance of 1 m with 49% of these being classed as serious. By comparison, the figures for security radio handsets were 35% and 49%, respectively, and those for mobile phones 4% and 0.1%.[3] It is important to note that these devices can cause interference even in standby mode and must, therefore, be fully switched off to be considered safe. Cordless telephones and wireless computer local area networks do not appear to cause significant interference.

Although most hospitals have policies demanding that mobile phones be switched off in clinical areas, clinicians must always bear in mind the potential for such malfunction, given the ubiquitous nature of mobile telephones and the increased risk of problems with the two-way radios used in hospitals.

TARGET-CONTROLLED INFUSION (TCI)

Intravenous anaesthesia, referring to anaesthesia maintained by continuous infusion of anaesthetic agents, requires that consistent drug concentrations can be achieved which may be altered in response to the needs of the procedure. The term TIVA is used synonymously with intravenously maintained anaesthesia, but perhaps should be reserved for those scenarios where no inhalational agents (including nitrous oxide) are co-administered.

Although the drugs used for intravenously maintained anaesthesia are of rapid onset and have short half-lives and durations of action, in order to rapidly achieve and maintain a given clinical effect it is still necessary with currently available agents to administer boluses and then to reduce progressively the infusion rate. TCI describes a system whereby a computer controls the rate of infusion of a drug to achieve (in as short a time as possible) and maintain any given target concentration. In use this means that the anaesthetist is relieved of having to continuously make complex calculations and adjustments of the infusion rate, thus lending the ability to endlessly vary the drug concentration to achieve the desired effect (Fig. 19.14).

Investigators in the fields of anaesthesia and pharmacokinetics have considerable experience in programming computers to achieve constant concentrations of drug by driving infusion pumps at variable rates. Many different terms have been used to describe this process since the description of the bolus elimination and transfer (BET) infusion scheme with a system called CATIA (computer assisted total intravenous anaesthesia) in 1983.[4] The aim throughout has been to emulate the simplicity of administration of inhalational agents, in effect to design a 'calibrated vaporizer' for intravenous agents, i.e. a system akin to that for inhalational agents wherein a given setting on the 'vaporizer' will, within certain limits, result in a similar and constant drug concentration in the patient.

Although the original experimental systems were cumbersome, in 1990 workers from the University of Glasgow described a system for the delivery of propofol using a conventional syringe pump driven by a Psion II hand-held computer.[5] There was good correlation between measured and predicted plasma concentrations, and users found the

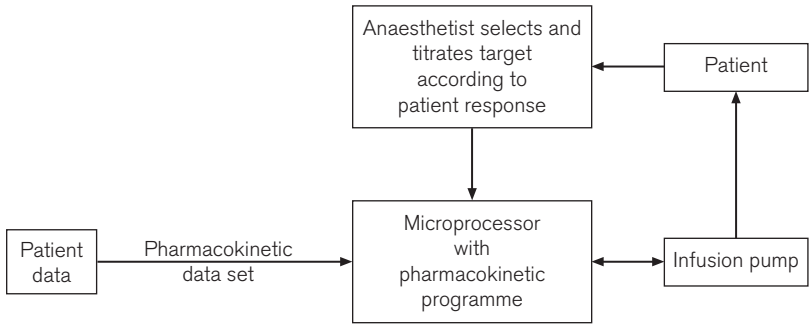


Figure 19.14 Relationships in a target-controlled infusion system.

device easy to use. This was the basis for the Diprifusor system, for many years the only commercial system for delivering a drug using a pharmacokinetic algorithm. Since 2005 various device manufacturers have incorporated TCI algorithms into their infusion pumps, these are examined under 'open label TCI'.

Key components

A TCI system must have the following key components:

- an infusion device
- a computer, and a pharmacokinetic algorithm for each drug which translates predictions from the model into instructions to control the infusion device, the algorithm must have the correct pharmacokinetic data set (constants) for the patient group in which it will be used
- patient-specific data: usually weight (the Diprifusor system does not alter the data set for patient age or circumstance); some algorithms may need other co-parameters such as age, sex and height
- a user interface to allow and confirm data input and to inform the user of the algorithm's predictions
- a fail-safe system: a safety mechanism to shut down the system in the event of computer failure; this may be in the form of a second microprocessor calculating the drug concentration based upon the output of the pump and comparing this with the predicted value from the first microprocessor's control algorithm, with instructions to shut down the system if there is greater than a preset discrepancy (5% for Diprifusor).

It is important that the algorithm is not inadvertently reset as if for a new patient when renewing drug syringes during the course of an anaesthetic, as this will of course result in the delivery of a further loading dose.

Diprifusor

In 1996 the manufacturer Graseby produced anaesthesia syringe drivers bearing the 'Diprifusor' TCI Subsystem

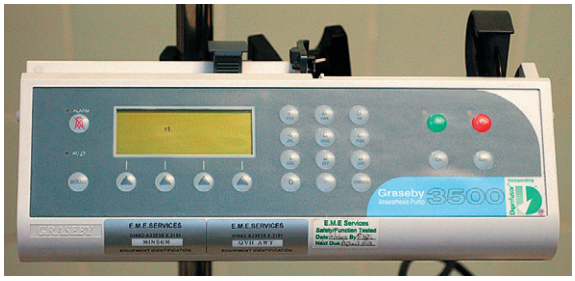


Figure 19.15 Graseby 3500 Anaesthesia Pump incorporating the Diprifusor chip and demonstrating the associated logo.

(Fig. 19.15) licensed from Zeneca (now AstraZeneca), the makers of propofol. The Diprifusor chip has subsequently been incorporated into other makes of syringe driver, the devices being readily identified by the associated green and white logo. An advantage of this modular approach to the design of TCI systems, where the pharmaceutical company retains responsibility for the Diprifusor module, is the standardization of propofol delivery, such that all pumps bearing that module will deliver the same amount of drug for any given target setting because the same pharmacokinetic model is operating (cf. open label TCI issues). To date, in excess of 20 000 such modules (Fig. 19.16) have been sold worldwide (personal communication, AstraZeneca). The Subsystem comprises components designed to:

- recognize electronically tagged pre-filled syringes of propofol produced by AstraZeneca in order to identify drug and concentration and enable TCI mode when appropriate
- control syringe pump in TCI mode to deliver propofol by TCI.

In order to ensure that the syringe driver operates in TCI mode only when loaded with the correct drug (proprietary propofol as Diprivan), the pre-filled syringes carry a small electronic tag in the finger grip of the barrel. When the syringe is correctly located the 'Programmable Magnetic

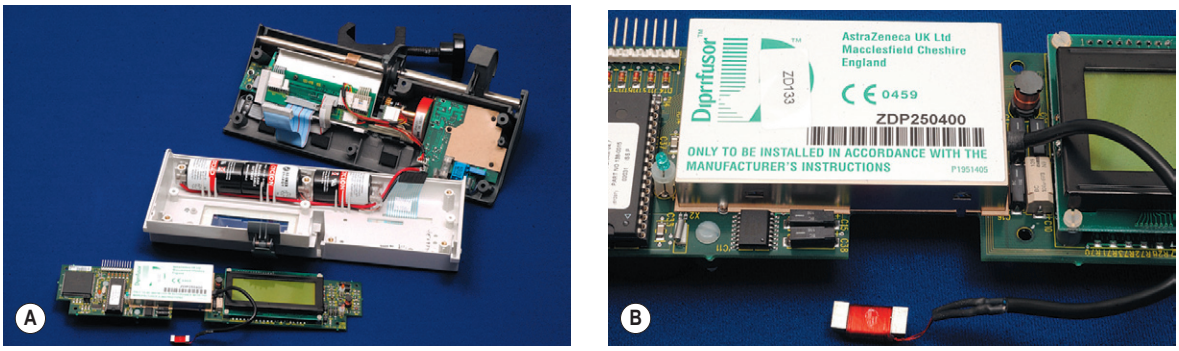


Figure 19.16 **A.** The Diprifusor subsystem installed on the microprocessor control board of the Graseby 3500. **B.** Note the 'aerial' (A) for recognition of the syringe tag.

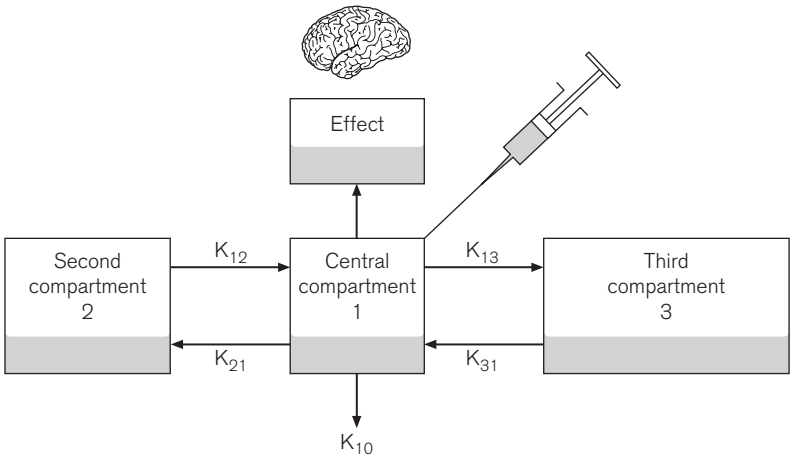


Figure 19.17 Schematic representation of open three-compartment model.

Resonance' tag lies within a recess close to an aerial in the body of the pump. In response to signals from the aerial the magnets within the tag oscillate at particular frequencies generating a signal, which is picked up by the aerial. To discourage refilling of the syringe the tag is subsequently 'wiped' by the pump when the actuator pushing the syringe plunger reaches a pre-set travel.

The pharmacokinetic model used is based on the Marsh open three-compartment model[6] (Fig. 19.17) with parameters as shown in Table 19.1. Fig. 19.18 shows a typical infusion scheme from a TCI device to demonstrate the bolus and decreasing infusion rate needed to maintain an increased concentration. Diprifusor software (which targets the plasma concentration as the control variable) also allows the predicted effect site concentration to be demonstrated at any time. This facility was added slightly later using a rate constant for equilibration of the effect site (k_{e0}) of 0.26 min^{-1} derived from a separate study.[7]

Table 19.1 Pharmacokinetic parameters for propofol incorporated in Diprifusor software

V_1 Volume of central compartment	228 ml kg^{-1}
Half life of delay central to effect site ($t_{1/2}$Keo)	2.6 min
Rate constants:	
K_{10} (elimination rate constant)	0.119 min^{-1}
K_{12}	0.114 min^{-1}
K_{21}	0.055 min^{-1}
K_{13}	0.042 min^{-1}
K_{31}	0.003 min^{-1}

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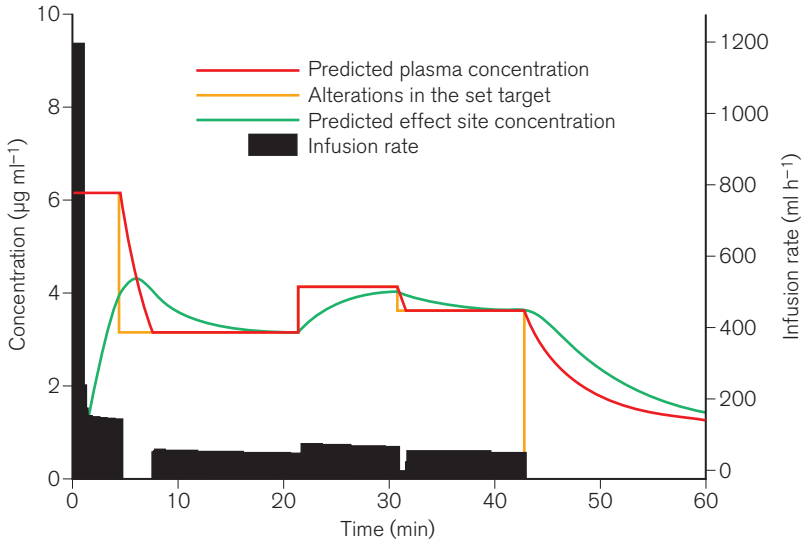


Figure 19.18 A typical infusion scheme from a target-controlled infusion device, showing the predicted plasma concentration in response to alterations in the set target and the infusion rate.

Effect site display is useful, both in terms of demonstrating for impatient anaesthetists the magnitude of the time lag between the compartments and for correlating clinical effect with theoretical concentration when plasma compartment 'overpressure' is used to speed up the onset of effect.

Accuracy

Two independent factors contribute to the overall accuracy of a TCI system:

1. Delivery performance of the infusion system: how well the syringe pump outputs the desired volume of drug
2. Predictive accuracy of the pharmacokinetic model.

Syringe pumps incorporating Diprifusor are required by the manufacturer to have a performance such that the infusion error at specific time points is within ±5% of the ideal volume.[8]

In considering figures for the predictive accuracy of pharmacokinetic models, a number of issues must be borne in mind:

- inter-patient pharmacokinetic variability must always be expected and the need to titrate drug target concentration to achieve the desired effect is inescapable
- performance is likely to vary with time and chosen target concentrations
- errors in blood sampling technique or variability in the measurement of blood concentrations; this may at times be due to inadequate mixing at high infusion rates
- in clinical assessments of accuracy the co-administration of other drugs may affect the handling of the target drug.

There is an attempt to standardize measurement of the accuracy of computer-controlled infusion systems.[9] *Performance error* expressed as a percentage is calculated as shown below:

$$\text{Performance error}\,(\%) = \frac{C_M - C_{Calc}}{C_{Calc}} \times 100$$

where C_M and C_{CALC} are the measured and calculated blood drug concentrations. Other indices are derived from this. The *median performance error (MDPE)* is a measure of the tendency of the system to over- or underestimate the measured blood concentration in a given patient or scenario. This is the degree of bias and has direction as well as value, thus if bias has a positive value the measured concentrations tend to be greater than predicted. The precision, or size, of the error is represented by the median absolute performance error (MDAPE). Divergence and wobble reflect on time-related changes in performance and intra-subject variability in performance respectively. Typical figures for MDPE of 16.2% and MDAPE of 24.1% were reported for Diprifusor in one study,[10] with measured concentrations tending to be higher, particularly after induction or an increase in target concentration and at higher targets. In context these figures compare favourably with the performance of inhalational anaesthesia, where after 15 min of isoflurane administration a ratio of 0.78 is reported between arterial and end-tidal partial pressures[11] with the concentration remaining 20% lower at 1 h. In this context there is an even greater difference between vaporizer setting and arterial concentration.[12]

Open label TCI

The expiry of the patent on propofol and the general acceptance of TCI as a mode of drug delivery have driven other manufacturers to incorporate TCI algorithms into their syringe drivers. These are known as '*open label protocols*' because they do not require the use of the tagged syringes of Diprivan. At the same time the popularity of remifentanil as the opioid agent in TIVA naturally led to demands for an appropriate TCI algorithm for that drug also. The resulting devices have been approved under medical device legislation without any reference to the pharmaceutical company responsible for drug labelling, although the joint role of some manufacturers as being also producers of the drug propofol does appear to constrain their readiness to offer further models for propofol.

Currently in the UK, open label pumps are available from four manufacturers. All offer a choice of Schnider[13] and Marsh pharmacokinetic (PK) models to drive the propofol TCI and the same single model (Minto[14]) for TCI remifentanil.

TCI sufentanil using the Gepts[15] model is also available from all the device manufacturers mentioned here except B Braun. At the time of writing, two companies offer further drug models. CareFusion in the Alaris PK pump also offer paediatric propofol pharmacokinetic models – Kataria[16] and Paedfusor[17] – and a model for alfentanil (Maitre[18]). Arcomed AG (Switzerland) can offer two versions of the Marsh model as well as Kataria and Maitre models and have expressed a willingness to consider implementing other models (with due licensing disclaimers) at customers' specific requests (personal communication). B Braun and Arcomed AG are able to implement the TCI algorithms into their peristaltic infusion pumps as well as the more traditional syringe pumps, adding another degree of flexibility.

Where the pharmacokinetic (PK) models have the facility to calculate the effect site concentration the new open label pumps can offer the option of using the calculated effect site, rather than the plasma concentration, as the control variable for driving the infusion. Remifentanil effect site control using the Minto model is offered by all devices, similarly propofol effect site control on the Schnider model. Elsewhere calculations and control of effect site concentration have introduced a whole new area of confusion for users of open label pumps. The reasons for this, namely the methodological details of pharmacokinetic/pharmacodynamic (PK/PD) modelling and attendant controversies, are well beyond the remit of a text on equipment, but the consequences are relevant, as described below.

Ultimately the various devices all offer much the same functionality whether they are controlled through one interface as in the Orchestra Base Primea from Fresenius (Fig. 19.19) or programmed and controlled individually as in the Alaris (Fig. 19.20) and B Braun (Fig. 19.21)

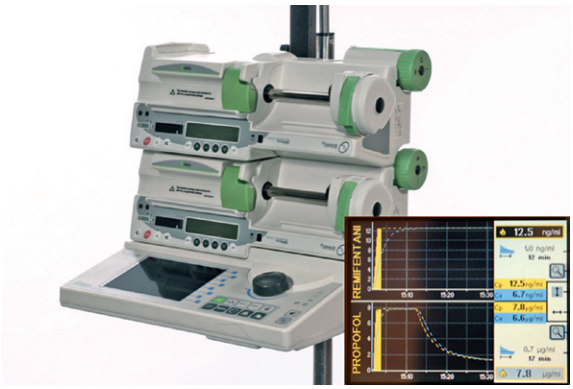


Figure 19.19 Orchestra Base Primea (Fresenius Kabi AG) allows simultaneous control of several devices through one control panel (screen display shown here as inlay).



Figure 19.20 Alaris PK pump.
Photograph courtesy of Carefusion.

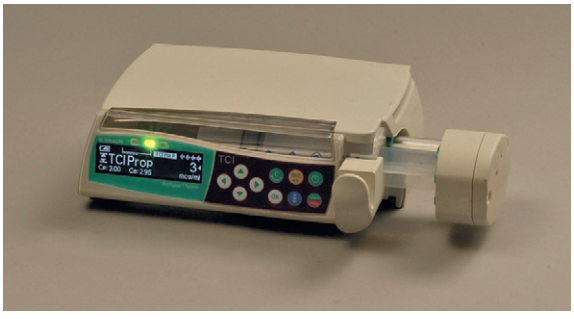


Figure 19.21 Perfusor Space, B Braun.

devices. For most prospective users the choice between them is probably more a matter of balancing personal preference for the input method and display against the irritating interface and programming quirks. The devices all suffer from not having a numeric keyboard – thus having to scroll through numbers for entering values – and needing preloading with infusion schemes to allow use in anything other than TCI mode or simple ml/h control.

The potential for confusion and error on the whole has been markedly increased. This is as a consequence of:

- having two or more pumps side by side with no automatic drug identification
- differing user interfaces and set up procedures between devices
- issues with pharmacokinetic models: their limitations and varied implementations by manufacturers, as below.

Other pharmacokinetic models

It is paramount for anaesthetists to recognize that a given target concentration obtained using one particular model (for example the Marsh model as used by Diprifusor) does not necessarily translate to the same clinical effect when that target is delivered by a different PK model (for example, the Schnider model as available in the open label TCI pumps). This is because it is precisely that which causes the models to be different (i.e. the predicted concentration following a given dose) that will cause them to give differing amounts of drug for a given target concentration. Clinical effect is clearly a function of dose of drug given and, hence, differing effects must be expected when using different models targeting the same concentration.

Additional confusion surrounds the issue of effect site control of the infusion, with multiple models and manufacturers updating the Marsh model with their own interpretations of available data (Table 19.2). The rate constant (k_{e0}) that was originally added to Diprifusor to calculate and display the effect site concentration of propofol was felt by some to be too slow, but because of the predominance of Diprifusor it is a figure that has clinical relevance for many and is used elsewhere. Fresenius uses a Marsh model with a much a larger (k_{e0}) for effect site control, but this is felt now to be too fast.[19] Arcomed offer two versions of Marsh for effect site control, each effectively using figures at opposite ends of the spectrum. The Alaris PK uses the same value as Diprifusor, and Braun use an intermediate figure for their Marsh model, but neither currently allows effect site TCI using the Marsh model.

Thus, in *effect site targeting* mode a given target concentration can result in different volumes of drug being delivered from different manufacturers even when nominally using the same model (see Fig. 19.30). Even for those with an interest in these issues, it is difficult to learn or remember how a PK model and device combination will behave under different circumstances, particularly once patient variability is also considered!

Extremes of weight also present a problem for the current PK models. This is most notable in the obese, where the James equations[20] used by the Schnider and Minto models in the calculation of lean body mass (LBM) for their algorithms, generate decreasing values above a certain body weight depending on height and gender. As a result Fresenius pumps will not progress with TCI for body mass index (BMI) >42 for males and >35 for females, these being the inflection points on the James equation graphs after which the LBM starts to decrease. Alaris pumps handle the problem differently; they simply will not allow a greater weight entry then the corresponding figure for the above BMIs. The Marsh model (and hence Diprifusor) markedly underestimates plasma concentrations for morbidly obese patients (i.e. delivers too much drug for a given concentration) and the patient weight used in the algorithm needs to be manipulated significantly towards the ideal body weight of 70 kg for both the obese and the underweight.

Future developments

Open TCI initiative

It can be seen from the problems above that great confusion can arise from the existence of the different models and their various implementations by the manufacturers. Significant detailed knowledge of the models and manufacturers' implementations of them is needed if anaesthetists are not to be surprised by these devices in use.

The 'Open TCI initiative' is an attempt by workers in the field of drug pharmacokinetics and pharmacodynamics to come up with a single unifying model for each drug on which all workers are agreed and which would be the basis for all dosing by TCI.[19] Manufacturers of medical devices would, thus, not be in the position of having to dictate drug dosing. It will also become possible to compare effects and outcomes between studies once drug delivery is based on a common dosing regimen.

Table 19.2 The differing rate constants for equilibration of the effect site with the plasma propofol concentration (also expressed as half time) used by different manufacturers

MODEL	MANUFACTURER	K_{E0} (MIN^{-1})	HALF TIME (MIN)
Marsh	Diprifusor	0.26	2.6
Marsh	Alaris	0.26	2.6
	B Braun	0.63	1.1
	Fresenius	1.2	0.57
Schnider	all	0.456	1.5

Experimental systems

A logical development from the TCI system, for maintaining constant concentrations of anaesthetic, has been to link this with some form of depth of anaesthesia monitor to automatically alter the chosen target to maintain a constant depth of anaesthesia. Closed loop anaesthesia (CLAN) has been successfully used to anaesthetize patients breathing spontaneously during surgery.[21,22]

TCI may also be used as the basis for an advanced form of PCA (or sedation). Here, the patient's button presses may, for example, result in an increase or prevent a decrease in the target concentration of drug rather than a drug bolus.[23] Such systems are so far experimental only.

PATIENT-CONTROLLED ANALGESIA (PCA)

The quality of postoperative pain relief and patients' sense of autonomy have perhaps improved with the use of PCA. Although intramuscular bolusing with opiates at 3–4 hourly intervals may provide good analgesia, the reluctance of patients to disturb hard-pressed nurses and the need for the increasingly sparse presence of a second member of staff to administer 'controlled drugs' rarely results in optimal analgesia. The advantages of PCA are:

- patient autonomy
- rapid relief of pain
- analgesia/dosage tailored to patient's requirements, with the patient able to balance analgesia and side-effects.

For safety reasons most physicians are reluctant to use a background infusion alongside the patient-controlled bolus doses. This is the key disadvantage: bolus doses are necessarily small with a short duration of action resulting again in fluctuating analgesia levels, particularly at night time where patients often complain of waking in pain and thus poor sleep.

Key points in any PCA system are:

- route of administration
- type of administration (bolus, background infusion)
- ease of programming (dose, lock out)
- ease of priming
- power source
- safety
- security
- portability
- display
- printout.

PCA is most commonly delivered intravenously, but may also be subcutaneous or into the epidural space, although epidural PCA does have the theoretical problem of bypassing the safety feedback loop seen with intravenous opiates, where the potentially overdosed patient becomes somnolent and unable to make further analgesic demands. The simplest method of administration is by bolus dosing; here a preset bolus, which may be programmed either by volume or in more complex systems by weight of drug in milligrams, is injected on demand. There is a pre-programmed lock-out time during which further demands are ignored, thus allowing for the time of onset of action of the previous bolus and protecting against overdose.

A background continuous infusion may also be available with some devices. This militates against patients waking in pain due to protracted periods of no analgesic administration, although safety considerations make this a rarely used facility (see above).

PCA devices (Fig. 19.22) are mostly variations of the infusion devices discussed earlier. They may be syringe drivers or peristaltic type pumps, which are microprocessor-controlled; hence, again the aspects of those devices mentioned elsewhere are still pertinent. Because of the trend towards increased patient mobility and patient-controlled epidural analgesia (PCEA), a number of manufacturers now produce miniaturized PCA pumps, often using a small compressible pumping chamber designed as an integral part of the giving set (Fig. 19.23). Size and portability



Figure 19.22 A Graseby PCA.



Figure 19.23 A Baxter PCA pump which may be used for epidural continuous infusion or even epidural PCA.

are limited by the size of the drug reservoir and the need to encase this in a lockable anti-tamper shell.

As with other microprocessor-controlled devices, there is a vast amount of information recorded by these newer devices, only some of which is presented to the clinician. This includes times of error messages, alarm conditions, 'power up' and 'power down' and rate or programme changes. There is usually a facility to view and/or download or print the 'patient history' so that pump settings may be tailored to the patient's needs. A further feature is the lockable control panel (this may be an actual key and lock in some machines) needing access codes and button presses of unlabelled 'soft-keys' to prevent unauthorized tampering with the pumps.

A consequence of increasing functionality, together with miniaturization and the need for security, is that many of these devices are no longer intuitive to use and effectively need a dedicated 'pilot'. Disenfranchised staff become reluctant or unable to initiate and alter settings with the quality of analgesia suffering as a result. Uniformity of equipment in clinical areas helps, but when deciding a purchasing policy, the considerable capital costs of these devices, the cost of initiation and maintenance of staff training, and the sheer numbers of devices needed to provide the service are also to be considered.

Elastomeric pumps

Given the issues surrounding microprocessor-controlled infusion pumps and the specific problems associated with the PCA configurations, there is much to be said for the use of simple low-cost disposable PCA devices, as shown in Fig. 19.24.

Elastomeric devices are in effect powered by the energy stored in the stretching of the balloon holding the reservoir of drug. A small compressible chamber of preset volume (usually 0.5 or 1 ml), on a wrist strap or attached to the reservoir, holds the bolus dose which is delivered to the patient through a one-way valve when the chamber is squeezed. The lock-out time is predetermined by a flow restrictor connecting the reservoir and chamber, which governs the rate of refilling of the chamber, thus giving a maximal flow rate through the device, typically 5 ml h^{-1}. Drug demands, before the effective lock-out time is reached, result in proportionately smaller doses. PCA regimen alterations are, therefore, made by altering the drug concentration in the reservoir. Some newer models have a user (prescriber) variable flow restrictor and or bolus size (Fig. 19.25).

Additionally, all such elastomeric devices if appropriately configured, can just as easily be used for the infusion of local anaesthetics either into the operative site or around major nerves for regional anaesthesia. For those devices designed for local anaesthetic infusion, the flow restrictor may be the catheter itself by virtue of its bore and length, with some systems allowing multiple such catheters to be run from one reservoir (e.g. Alpha range of pumps and catheters from Advanced Infusion, San Dimas, USA).

Flow accuracy for elastomeric devices is typically within 10–20% of the given rate, but depending on the design of the device, over- or under-filling of the reservoir can affect flow rate. Flow rate is also affected by temperature, particularly at the flow restrictor, and it is important that manufacturers' instructions are followed as to whether, for example, the tubing and restrictor are placed close to the patient's skin or worn outside the clothing. A change of 10°C in the temperature of water-based fluids results in altered viscosity causing a 20–30% change in flow rate.

In spite of these limitations, these devices are preferred by patients and nurses for their ease of use and, possibly, for the lack of alarms.

Of prime importance with PCA is safety and security. All devices must be safe against over-dosage, either caused by fault conditions or drug siphoning. There must also be security against tampering and adjustment of settings by unauthorized staff, patients or visitors.

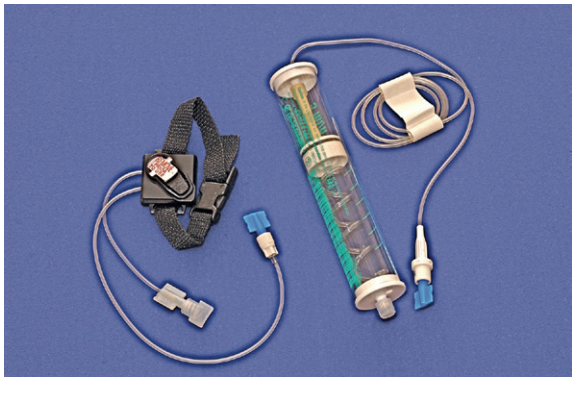


Figure 19.24 An elastomeric disposable PCA pump with the demand button worn as a 'wrist watch'.



Figure 19.25 ON-Q elastomeric pump with variable rate infusion controller and bolus device.
Photograph courtesy of I-Flow Corporation, Lake Forest, USA.

Other non-electrically powered infusion devices

Gas- and spring-powered devices do exist, particularly for specialist applications such as implanted drug-delivery systems. They are not discussed further here and are not in common usage for PCA or routine postoperative ambulatory analgesia.

AUTOTRANSFUSION

When a large blood loss may be expected during surgery, it may be possible to reuse the patient's own blood, provided that it is uncontaminated and free from clots. Uncontaminated blood is aspirated by the machine, red cells are washed, spun down and suspended in crystalloid solution with added anticoagulant and stored temporarily in a reservoir. It may then be transfused back into the patient via a filter. The operation of such machines needs a dedicated member of staff.

RELATED EQUIPMENT

Filtration

Intravenous fluids should be filtered to protect the patient from microscopic foreign material. Blood for transfusion that is more than 24 h old should be filtered (Fig. 19.26) to remove micro-aggregates that form from the breakdown products of the cellular components and platelets. Blood filters are of two basic forms: screen filters and depth filters. Screen filters function as 'sieves' and are usually constructed from a woven mesh. They have a regular pore size, often of 40 µm. The efficiency of a screen-type filter at removing foreign matter increases progressively with each unit of blood passed as the pore size tends to decrease progressively down to 20 µm. Screen-type filters are said to be less damaging to the red cells than are depth filters. Depth filters consist of a pack of synthetic fibre, often Dacron, not formed into a mesh. The mechanism of filtration is actually by adsorption of unwanted material, down to a size of about 10 µm onto the surface of the fibres. This adsorption is probably due to electrical charge differences between the particles and the fibres. With the depth filter, the efficiency at removal of unwanted material decreases with each unit of blood, probably as a result of channelling in the pack of fibres.

Intravenous crystalloids may be filtered with much finer sieve-type filters to remove foreign particulate matter, including bacteria. The Pall intravenous 'Site Saver' extends the life of the giving set, which should normally be replaced every 24 h, to up to 96 h. The construction of the 'Site Saver' is shown in Fig. 19.27. All particulate matter larger than 0.2 µm is removed by the 0.2 µm filter membrane. Air can be vented through a hydrophobic membrane, which has 0.02 µm pores.

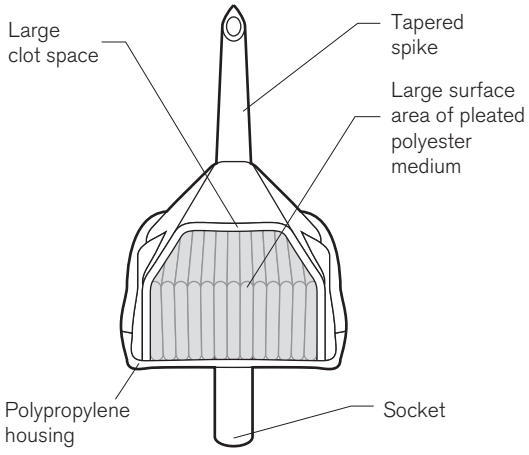


Figure 19.26 A blood filter.

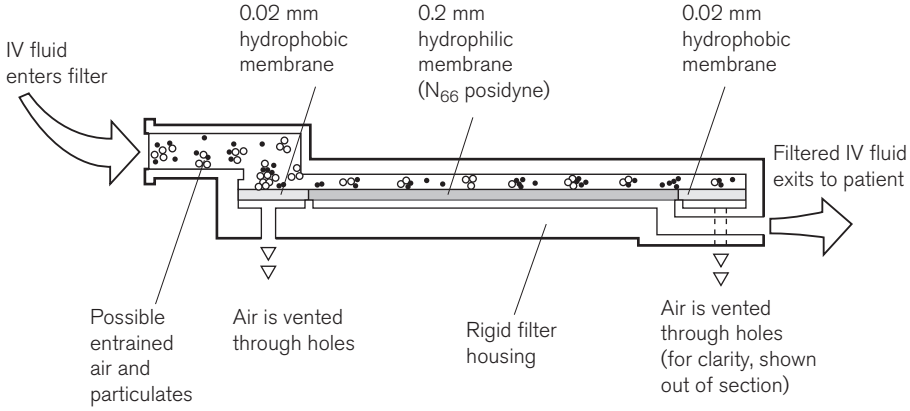


Figure 19.27 Pall intravenous 'Site Saver'.

Infusion lines

When more than one fluid or drug is infused through a common intravascular device, alterations in the rate of infusion of one of these substances will temporarily affect the rate of administration of the remainder. The degree of perturbation is a function of the shared volume (dead-space) of intravenous device and the infusion rates (Fig. 19.28). For drugs with short half-lives this can have a dramatic effect. Ideally, each intravenous infusion of drug would be given through a dedicated infusion line, so that alterations in the rate of infusion of one drug do not affect the others. For simplicity and patient comfort, it is preferable to insert only one intravenous catheter and use a many-tailed infusion set. Such infusion sets are available specifically for TIVA with one or two long narrow calibre 'pump lines' and one wider bore line to connect to a giving set for fluid administration under gravity (Fig. 19.29). Each line should incorporate a one-way or anti-reflux valve to prevent backflow as well as anti-siphon valves on the pump lines. The shared volume, where the lines join before the intravenous catheter, should be as small as possible. Because such lines are made up of a number of components assembled together it should be borne in mind that they may leak or be obstructed at the bonded junctions. This may not always be obvious.

There were previously the beginnings of a move towards yellow-coloured infusion lines for local anaesthetic applications; this is probably in abeyance now following recent developments. In response to previous well-publicized cases of maladministration of intrathecal cytotoxics and intravenous local anaesthetics, the National Patient Safety Agency of the UK issued an alert, in November 2009, requiring all UK hospitals to introduce syringes and needles with specific connections aiming to prevent '*wrong route*' errors. In spite of the commercial absence of such devices at present, it is required that, by April 2011, when intrathecal access is required this will be achieved using systems that do not connect with intravenous devices, i.e. will not be compatible with the ubiquitous Luer connectors.[24] A second part to the alert requires that this system is extended to include all epidural and other regional anaesthesia devices, 'but not local anaesthetics for small

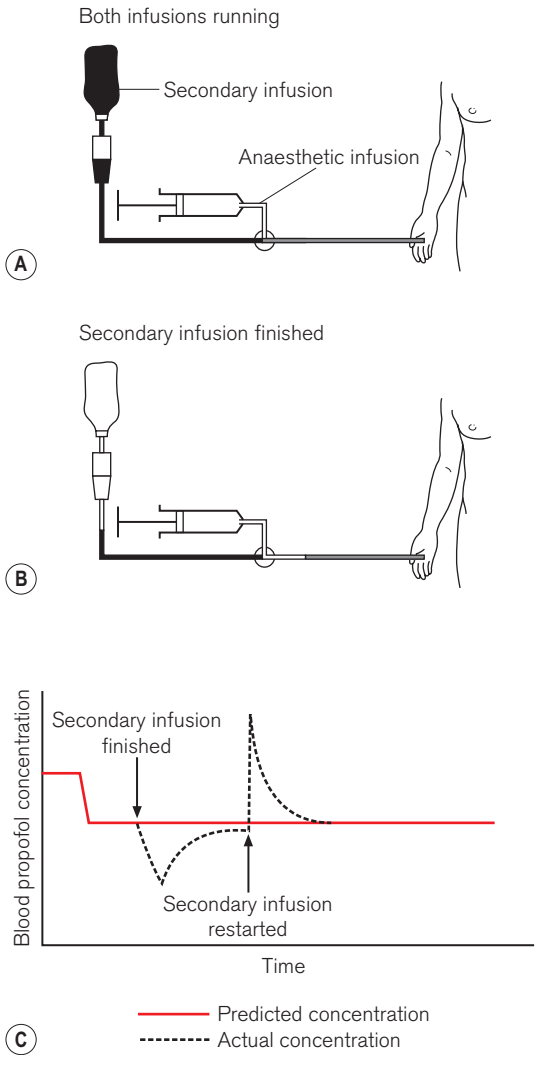


Figure 19.28 Changes in the flow rate of a secondary infusion influence the rate at which a drug is delivered to the body. For an infusion line with a volume of 5 ml between the connection of two infusions and the patient, if one infusion stops a considerable drop in drug concentration of the second may be expected. After restarting that infusion a bolus of the second drug is immediately administered, causing a peak in drug concentration.
Redrawn with permission from Engbers F, Vuyk J (1996) Anaesthesia Rounds, Target-controlled Infusion. *© The Medicine Group (Education) Ltd.*

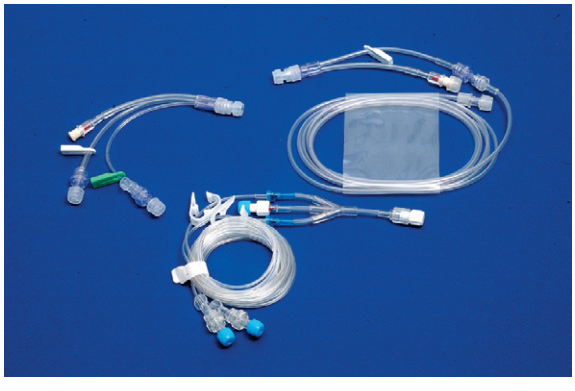


Figure 19.29 A selection of giving sets for intravenous anaesthesia (see text).

parts of the body', by April 2013.[25] Quite apart from the unknown risks and the resource implications, and the mayhem that is likely to result from needing a duplicate system of syringes and needles available in clinical areas, it is also untested and unproven whether this approach will ultimately reduce the, thankfully, small number of significant wrong-route errors.

TIVAtrainer©

For the purpose of understanding and predicting the disposition of intravenous drugs, no amount of theoretical knowledge of pharmacokinetics or practical experience of intravenous anaesthesia is as effective as seeing graphical representations of drug concentrations and the changes in response to interventions. TIVAtrainer© (copyright F. Engbers, Leiden University Hospital) is a pharmacokinetic simulation programme for intravenous anaesthetics developed as shareware available from the web site for the European Society for Intravenous Anaesthesia – EuroSIVA. TIVAtrainer (Fig. 19.30) is not the first simulation software of this type,[26] but it is perhaps the most developed and 'user friendly'. It is an ideal teaching tool for use in the operating room or the lecture theatre, allowing simultaneous modelling of several different drugs used in anaesthesia.

The programme can simulate manual and TCI infusion schemes demonstrating the infusion rates and associated plasma and effect site concentrations, as well as representing the amount of drug in each of the model's theoretical compartments, which are drawn to scale. A number of additional features permit the calculation of drug cost, the demonstration of Context Sensitive Half-Time and the hypnotic interactions between opiates and anaesthetic agents. The 'Help' file is well referenced and the programme provides a comprehensive teaching package. As with all mathematical modelling, extrapolations at the limits of the model must be interpreted with particular caution.

In use, a drug is chosen from the menu and a method of administration is chosen, which may be:

- manual bolus and infusion rates
- TCI with targeting of either the plasma or effect site compartments
- a further mode called 'IV assist' is designed to present simplified stepped infusion rates for entry on a manual infusion pump in order to mimic TCI.



Figure 19.30 A screenshot from Tivatrainer© showing three simulations of drug delivery to a target of 4 μg ml^{-1} in the same 70 kg patient. The numbers in black show the total volumes delivered at about 2 minutes. **A.** A device using the Marsh model with a k_{e0} of 0.26 min^{-1} in effect site control mode; **B.** a Fresenius device using the Marsh model in effect site control mode; **C.** any device using the Schnider model with effect site targeting mode. The white bars represent the infusion rate. For an explanation of the different volumes delivered see text.

REFERENCES

1. Medical Devices Agency. *Infusion Systems, Device Bulletin. MDA DB 2003(02)*. London: Medical Devices Agency; 2003.
2. Medical Devices Agency. *MHRA DB 2010(03)*. London: Medical Devices Agency; 2010. Available online at www.mhra.gov.uk
3. Medical Devices Agency. *Electromagnetic Compatibility of Medical Devices with Mobile Communications. DB9702*. London: Medical Devices Agency; 1997.
4. Schüttler J, Schwilden H, Stoeckel H. Pharmacokinetics as applied to total intravenous anaesthesia. *Anaesthesia* 1983;**38**(suppl):53–6.
5. Kenny GNC, White M. A portable computerized control system for propofol infusion. *Anaesthesia* 1990;**45**:692–3.
6. Marsh B, White M, Morton N, Kenny GN. Pharmacokinetic model driven infusion in children. *Br J Anaesth* 1991;**67**:41–8.
7. White M, Engbers FHM, Schenkels MJ, Burm AGL, Bovill JG. The pharmacodynamics of propofol determined by auditory evoked potientials. Abstract presented in Sydney at the *11th World Congress of Anaesthesiologists* 1996;A608.
8. Glen JB. The development of 'Diprifusor': a TCI system for propofol. *Anaesthesia* 1998;**53**(suppl 1):13–21.
9. Varvel JR, Donoho DL, Shafer SL. Measuring the performance of computer-controlled infusion pumps. *J Pharmacokinet Biopharm* 1992;**20**:63–94.
10. Swinhoe CF, Peacock JE, Glen JB, Reilly CS. Evaluation of the predictive performance of a 'Diprifusor' TCI system. *Anaesthesia* 1998;**53**(suppl 1):61–7.
11. Frei FJ, Zbinden AM, Thomson DA, Reider HU. Is the end-tidal partial pressure of isoflurane a good predictor of its arterial partial pressure? *Br J Anaesth* 1991;**66**:331–9.
12. Dwyer RC, Fee JPH, Howard PJ, Clarke RSJ. Arterial wash-in of halothane and isoflurane in young and elderly adult patients. *Br J Anaesth* 1991;**66**:572–9.
13. Schnider TW, Minto CF, Gambus PL, Andresen C, Goodale DB, Shafer SL, et al. The influence of method of administration and covariates on the pharmacokinetics of propofol in adult volunteers, *Anesthesiology* 1998;**88**:1170–82.
14. Minto CF, Schnider TW, Shafer SL. Pharmacokinetics and pharmacodynamics of remifentanil. II. Model application. *Anaesthesiology* 1997:**86**:24–33.
15. Gepts E, Shafer SL, Camu F, Stanski DR, Woestenborghs R, Van Peer A, et al. Linearity of pharmacokinetics and model estimation of sufentanil. *Anaesthesiology* 1995;**83**:1194–204.
16. Kataria BK, Ved SA, Nicodemus HF, Hoy GR, Lea D, Dubois MY, et al. The pharmacokinetics of propofol in children using three different data analysis approaches. *Anaesthesiology* 1994:**80**:104–22.
17. Absalom A, Amutike D, Lal A, White M, Kenny GN. Accuracy of the 'Paedfusor' in children undergoing cardiac surgery or catheterization. *Br J Anaesth* 2003;**91**:507–13.
18. Maitre PO, Vozeh S, Heykants J, Thomson DA, Stanski DR. Population pharmacokinetics of alfentanil:the average dose – plasma concentration relationship and interindividual variability in patients. *Anaesthesiology* 1987;**66**:3–12.
19. Glen JB. Minutes of the World Society of Intravenous Anaesthesia Open TCI Initiative, Copenhagen, 1 June 2008. http://opentci.org/doku.php?id=minutes:copenhagen accessed 15/7/10
20. James WPT. *Research on obesity*. London: Her Majesty's Stationery Office; 1976.
21. Kenny GN, Mantzaridis H. Closed-loop control of propofol anaesthesia. *Br J Anaesth* 1999;**83**:223–8.
22. Morley A, Derrick J, Mainland P, Lee BB, Short TG Closed loop control of anaesthesia: an assessment of the bispectral index as the target of control. *Anaesthesia* 2000;**55**:953–9.
23. Irwin MG, Jones RD, Visram AR, Kornberg JP. A patient's experience of a new postoperative patient-controlled analgesic technique. *European Journal of Anaesthesia* 1994;**11**:413–5.
24. National Patient Safety Agency. *Safer spinal (intrathecal) epidural and regional devices- Part A. 24 November 2009. NPSA/2009/PSA004A*. London: National Patient Safety Agency; 2009. Available at http://www.nrls.npsa.nhs.uk/alerts/?entryid45=65259

Update to reference 24, Safer spinal (intrathecal), epidural and regional devices – Part A update 31 January 2011. http://www.nrls.npsa.nhs.uk/resources/patient-safety-topics/medical-device-equipment/?entryid45=94529, accessed 28/7/11.

25. National Patient Safety Agency. *Safer spinal (intrathecal) epidural and regional devices- Part B. 24 November 2009. NPSA/2009/PSA004B*. London: National Patient Safety Agency; 2009. Available at http://www.nrls.npsa.nhs.uk/alerts/?entryid45=65259
26. Stanford PK/PD server: http://anesthesia.stanford.edu/pkpd/default.aspx, accessed 08/08/2010.

FURTHER READING

Absalom AR. *Struys MMRF. Overview of target-controlled infusions and total intravenous anaesthesia*. Gent: Academia Press; 2005.

Buchmann I. *Batteries in a portable world: a handbook on rechargeable batteries for non-engineers*. 2nd ed. Richmond: Cadex Electronics Inc.; 2010. Available at http://www.buchmann.ca/default.asp accessed 12/07/2010).

Medical Devices Agency. *Infusion Systems, Device Bulletin. MDA DB2003(02)*. London: Medical Devices Agency; 2003.

White PF, editor. *Textbook of intravenous anaesthesia*. Baltimore: Williams & Wilkins; 1997.

Chapter 20

Medical suction apparatus

Andrew J Davey

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Suction apparatus is vital to safe medical practice, especially in anaesthesia, resuscitation and intensive care. It is used for the clearance of mucus, blood and debris from the pharynx, trachea and main bronchi. During surgery, suction is used to provide a clear operating field for the surgeon. Specially adapted suction apparatus can also be used for other procedures, such as gastrointestinal, wound and pleural drainage.

MAIN COMPONENTS

The main components of a medical suction system are:

- energy source
- conversion of that energy to vacuum
- filter
- collection vessel.

The schematic drawings in Fig. 20.1 outline the methods used in suction apparatus, using internationally agreed symbols.

Energy source

Suction apparatus requires an energy source that generates a sub-atmospheric pressure. This is colloquially referred to as a vacuum source, despite the fact that a true (high) vacuum is rarely achieved. In fact, the pressure required is only a maximum of 60 kPa less than the normal environmental pressure (see Further reading).

The energy sources most commonly used are mains electricity and pipeline suction. Pipeline suction, of course, is a source of vacuum generated by an electrically powered vacuum pump at a distance from the user.

Portable suction apparatus may be battery driven, hand or foot operated, or may make use of compressed gas as a source of energy.

Vacuum source

The sub-atmospheric pressure required may be generated by:

- an electric motor or other source of rotational energy that may be used to drive a mechanical pump, various forms of which are shown in Figs 20.2A–E.
- pneumatically driven pumps that usually work on the Venturi principle (Fig. 20.2F). The driving force may be air, oxygen, steam or water
- a manually (or foot) operated spring-loaded bellows arrangement with unidirectional valves (Fig. 20.2G).

Pump types

Fig. 20.2A shows a piston pump, which is capable of creating high vacuum but, in transportable models, has a relatively low displacement (i.e. can only sustain low flow rates). Fig. 20.2B shows a diaphragm pump, which is a variation of the piston pump. It is mechanically simpler but is also frequently much noisier. One reason for the increased noise is that often, rather than using

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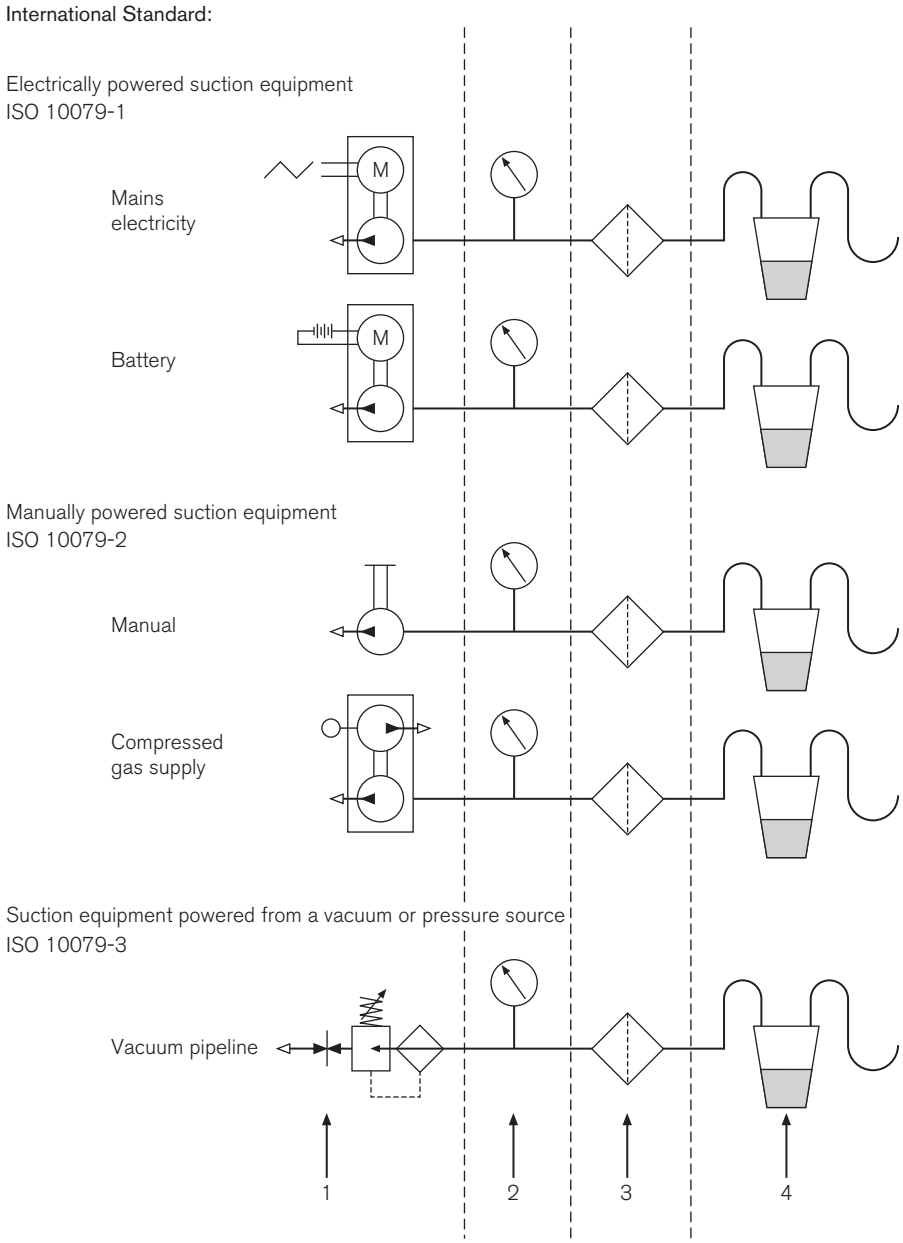


Figure 20.1 Suction equipment. Key: (1) vacuum source/regulator; (2) vacuum indicator; (3) filter; (4) collection container. (See ISO documents in Further reading.)

a conventional rotating electric motor, a much simpler large electromagnet displaces the diaphragm. Fig. 20.2C shows a form of rotary pump that can produce a high vacuum without conventional one-way valves. Fig. 20.2D shows a rotary pump capable of producing very high flows, as would be required in a dental surgery. It works in the same way as a vacuum cleaner and has the disadvantage of being extremely noisy. Fig. 20.2E shows a pump that works on the principle of the Archimedean screw. This type of pump can produce a high vacuum for a comparatively small size of machine. Fig. 20.2F shows the principle of a Venturi pump which makes use of the Bernoulli effect. Compressed fluid (gas or liquid), passing through a narrow orifice, creates a region of negative pressure

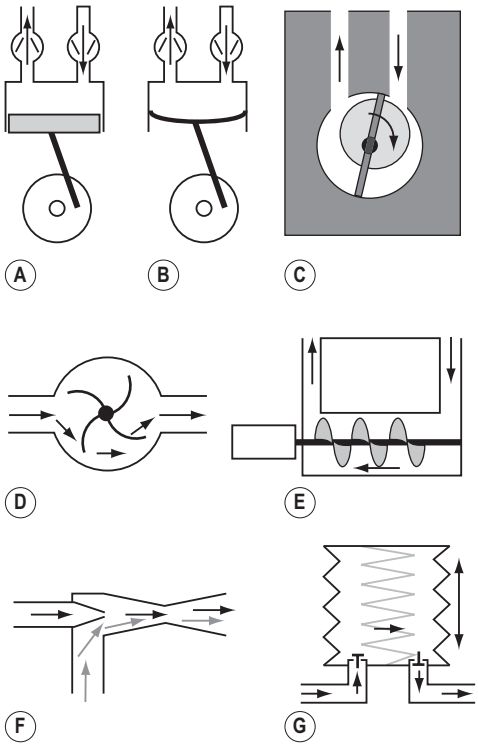


Figure 20.2 Vacuum pumps. **A.** Piston pump; **B.** diaphragm pump; **C.** high vacuum rotary pump; **D.** low vacuum rotary vane pump; **E.** rotary pump using an Archimedean wheel; **F.** gas-powered pump using the Venturi principle; **G.** a bellows pump.

beyond that orifice, which can be used to entrain adjacent air/debris. The main disadvantage of this simple affair is that it uses and is thus wasteful of large volumes of driving fluid, which is usually oxygen from cylinders. However, it does have the virtue of being extremely portable (Fig. 20.3). Finally, Fig. 20.2G shows a simple bellows mechanism with a pair of one-way valves, as would be used in manually operated suction apparatus.

The pump may be:

- permanently sited as in pipeline suction systems
- transportable, usually powered by mains electricity and supported on castor wheels; or
- truly portable, powered by battery, a cylinder of gas or by human energy (hand or foot operated).

There are international standards for each type of suction apparatus and these are listed at the end of this chapter.

Internal connections

The vacuum source is connected to the filter and collection vessel by tubing that is as rigid as possible to avoid collapse due to the vacuum. In pipeline systems the tubing can be made from metal or hard plastics, but the internal tubing in transportable units has to be firm but flexible, as does the final connection to the collection vessel in all types.

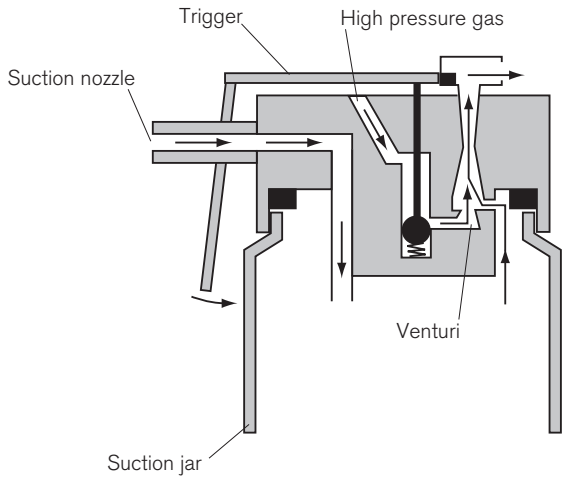


Figure 20.3 Working principles of a suction device using a Venturi injector.

Filter

This is fitted between the collection vessel and the vacuum source. It is to prevent contamination reaching the pipeline or pump and being expelled into the atmosphere. Filters should remain dry and must be replaced at the intervals recommended by the manufacturer, otherwise there is risk of reducing suction efficiency and of the filter becoming an infection risk itself.

Collection vessel

The collection vessel stores the aspirate prior to its disposal. It needs to be of sufficient volume to hold enough aspirate so that it does not need emptying too often. However, the larger its volume, the greater the displacement that is necessary from the pump, in order to reduce the pressure in the vessel before aspiration can occur efficiently. It is important that the inlet to the collection vessel has an internal diameter sufficiently large to admit the largest particles expected in the particular application of the apparatus (see Further Reading).

To minimize risks of contamination to theatre staff, all suction apparatus, except portable emergency units, should make use of a disposable collection system (Fig. 20.4A). This usually consists of a rigid outer transparent container with volume markings on its side and an inner (transparent) disposable plastic sleeve with an integral lid. The aspirate is stored in the inner sleeve and when full

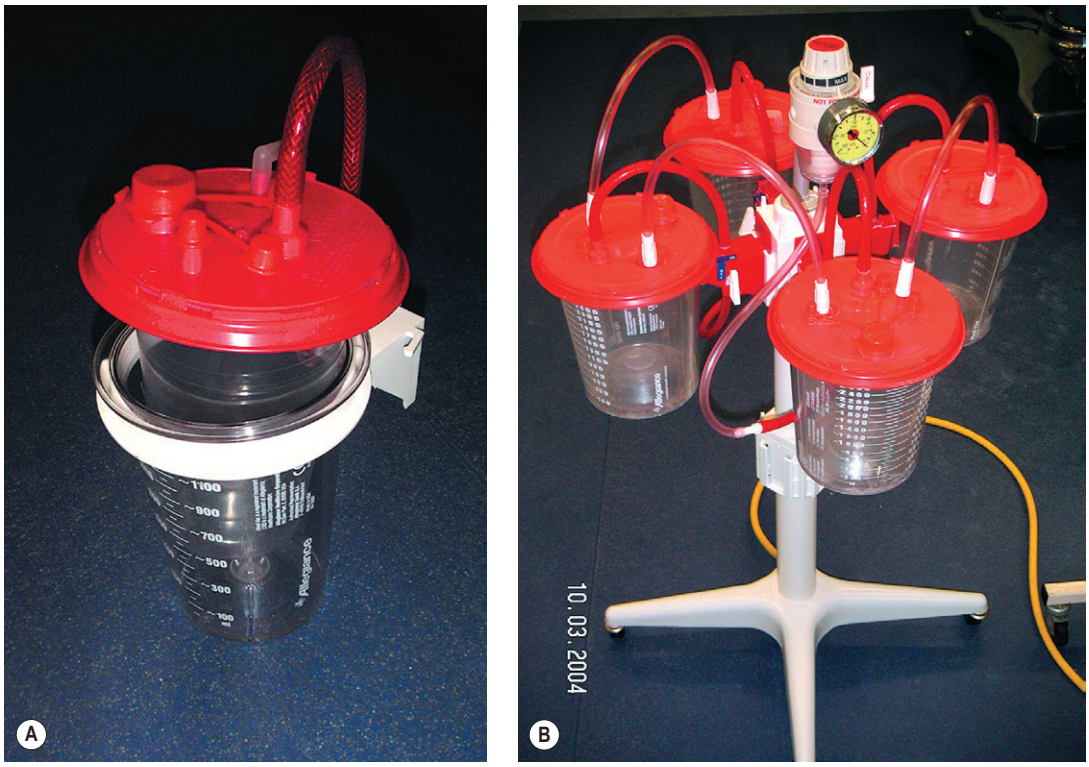


Figure 20.4 **A.** Disposable collection vessel system. Note the outer container with volume markings and the inner sleeve fitted with a lid. **B.** A disposable collection system with multiple collection vessels.

is capped off and discarded. As it is disposed of with other clinical waste, a gelling agent should be added to the vessel so as to solidify the contents to prevent accidental spillage. Thus, staff should never come into contact with any aspirate.

Suction tubing to disposal

There must be a suitable length of suction tubing, the patient end of which is usually fitted with a detachable rigid suction hand-piece. This tubing is a compromise between rigidity, to avoid collapse when the vacuum is applied, and flexibility, for ease of use. With portable hand-held apparatus, the hand-piece may be connected directly to the collection vessel for ease of use.

Efficiency

The efficiency of suction apparatus depends upon:

- the degree of vacuum (*sub-atmospheric pressure*) that can be produced by the pump, with particular regard to the time taken to achieve it
- the displacement, i.e. the volume of air at atmospheric pressure that the pump is able to move in unit time
- the internal resistance of the suction apparatus as a whole. This is related not only to the length and diameter of tubing and other components, but also to the tubing and other accessories between the apparatus and the liquid being aspirated
- the viscosity of the matter being aspirated.

Different pump designs may have differing displacements and high-vacuum capabilities and, therefore, may be selected for specific tasks. For example, liposuction requires high vacuum to dislodge fat globules that have a high viscosity but a small displacement because only a relatively small volume and no air are removed. In contrast, dental suction requires high displacement to remove large amounts of water spray, air and dental debris, but does not require a high vacuum. This type of apparatus must also have a low internal resistance and be connected to a relatively wide-bore hand-piece and tubing to maximize the rate of removal of debris.

OTHER COMPONENTS OF SUCTION APPARATUS

Vacuum control valve or regulator

This may be fitted between the vacuum source and the collection vessel. A vacuum control valve is a bleed valve that when opened, admits air, thereby reducing the degree of vacuum. A vacuum regulator operates on a similar principle to a pressure regulator and so no energy is wasted from the system. Regulators are always used with pipeline systems, whereas control valves are common with transportable electrically driven units.

Vacuum gauge

The vacuum gauge is also placed between the vacuum source and the collection vessel. It is calibrated in mmHg or kPa or both scales. The purpose of the gauge is three-fold: to test the apparatus for efficiency and leaks, to allow for adjustment of the available vacuum during use and to warn of excessive suction being applied to tissues. Note that modern vacuum gauges indicate counter-clockwise.

Cut-off over-flow valve

This is part of the collection vessel assembly and prevents any aspirate from leaving the vessel and entering the vacuum source or controller. It usually consists of a float, which is lifted by the rising level of aspirate when the vessel is full.

In order to test both the collection vessel and especially its inlet and also the cut-off overflow valve, an international standard test vomit has been specified. It consists of food-grade xantham gum, water and 1 mm glass beads (see Further reading).

Foam prevention

Foaming can be a problem in the collection vessel. Rising foam may cause premature closure of the overflow cut-off valve or it may fail to activate the valve and pass beyond it, contaminating the filter or pump and causing failure. Foaming also makes it difficult to estimate the amount of aspirate in the vessel. It may be reduced by the addition of a small quantity of silicon-based emulsion or by silicon coating of the inside of the collection vessel.

Multiple collection vessels

Modern suction apparatus usually has two collection vessels with a manually operated valve, so that the vacuum source can be rapidly switched from a full vessel to a fresh one. Alternatively, where large volumes of aspirate are anticipated, a number of collection vessels are connected in series in a carousel with all the individual valves open. Here, aspirate overflows from the first vessel into the next and so on (Fig. 20.4B). When the accurate estimation of small volumes of aspirate is required, as in paediatric surgery, a small calibrated vessel may be used in addition to the main apparatus. This container is disposable and is usually close to the operative field.

The suction nozzle, catheter or hand-piece

The design of what is referred to in International Standards as the 'applied part' depends upon the application. The commonest examples of hand-held suction nozzles are shown in Fig. 20.5. The key requirement of any suction 'applied part' is that the smallest internal diameter is at the very tip. If there are smaller diameters between the tip and the collection vessel, then blockages are likely to occur. The shape of the tip should be smooth so as to prevent damage to delicate tissues. The practice of allowing the tip to be occluded by any tissue, to reduce the noise of suction when not actually aspirating, must be deprecated as it causes tissue damage. Similarly, the noise of suction when using transportable, electrically powered suction machines should be reduced by switching off the motor rather than by occluding the suction tubing, as this may overload the motor. Hand-held suction nozzles of the Yankauer type (Fig. 20.5) often have a hole on the handle for fine control of suction with a finger.

Bronchial suction catheters should have smooth tips to avoid damaging delicate tissues and usually have several holes around the tip.

To protect staff, closed suction catheter systems are now commonly used (Fig. 20.6). The catheter is enclosed in a flexible sheath, which is permanently attached to a special 15 mm-taper adapter, which is left in circuit between the catheter mount and the endotracheal or tracheostomy tube. These systems are replaced every 24 h or as specified by the manufacturer.

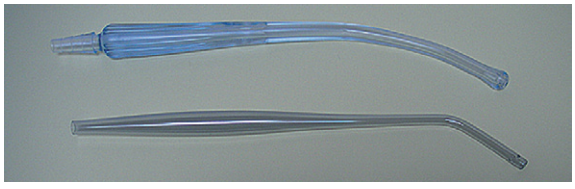


Figure 20.5 Two examples of hand-held Yankauer suction nozzles.

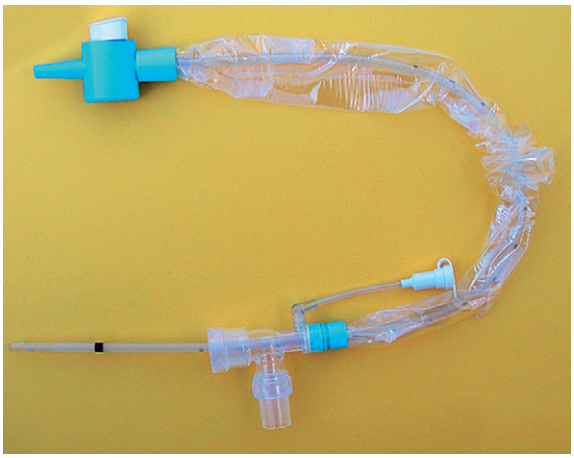


Figure 20.6 Closed bronchial suction system.

LOCAL VACUUM UNITS

Piped vacuum systems are now installed in most hospitals. The 'behind the wall' equipment and terminal outlets are described in Chapter 1. Medical suction apparatus may be connected into this system, wherever there is a terminal outlet.

There are two main types of 'local' apparatus:

1. Free-standing floor units, mains electricity powered and often with two collection vessels used for surgical purposes in the operating theatre (Fig. 20.7). Small, 'low-suction', low displacement units are also available for bedside use for intracavitary and continuous wound drainage. Generating typically 5–50 mmHg sub-atmospheric pressure, these devices are usually mains and rechargeable battery-powered piston pumps.
2. Wall-mounted units are local suction controllers for connection to central piped vacuum source (see Chapter 1). These have a single collection vessel as shown in Fig. 20.8. There are two types of controller: conventional 'high-suction' and also 'low-suction'. Low-suction controllers are deliberately limited to provide safe suction for intra-pleural drainage or nasogastric suction. Confusion between the two types of suction controller can have disastrous consequences, hence the resurgence of free-standing 'low-suction' units.

CHOICE OF SUCTION APPARATUS

When selecting a suction apparatus for a particular purpose, the following points need to be considered:



Figure 20.7 Free-standing medical suction unit.

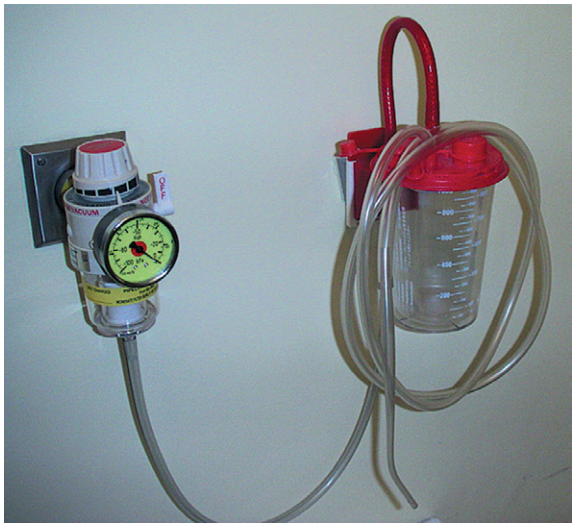


Figure 20.8 A pipeline vacuum unit. On the right is the controller, which is plugged directly into a flush-fitting outlet (obscured in this picture). On the right is the reservoir jar.

- Must it be portable? If so, should it be hand/foot operated (Fig. 20.9)? If not, should it be powered by electricity (mains or battery) or is pipe-line vacuum available? Could it be powered from a gas cylinder using a Venturi injector?
- Is a high displacement needed?
- Is a high vacuum needed?
- What size should the collection vessel be?



Figure 20.9 Ambu ResCue hand-operated suction apparatus. *Courtesy of Ambu International Ltd.*

It is important to ascertain, for an electrically driven suction machine, whether it is rated for continuous or intermittent use.

High-volume aspirators, as used in dental surgery, use a pump similar to that used in a vacuum cleaner (Fig. 20.10A). They usually have several suction tubes of different diameters for different applications (Fig. 20.10B). These tubes and their nozzles should never be obstructed, as a high flow of air is required to cool the motor, which would otherwise overheat. This type of high-flow suction is needed in dental surgery to aspirate the spray of water used to cool the tooth during high-speed drilling.

STANDARDS AND TESTING

As with all other medical equipment, there are published standards for design and manufacture. These standards, now, not only relate to safety, but also to function and use. Typical ranges of volume for the collection container for specific uses are listed, as are standards for levels of suction and flow.

Previous editions of this book suggested methods of testing equipment including suction apparatus; however, as all equipment has become more complex, only tests recommended by the manufacturers should be carried out.

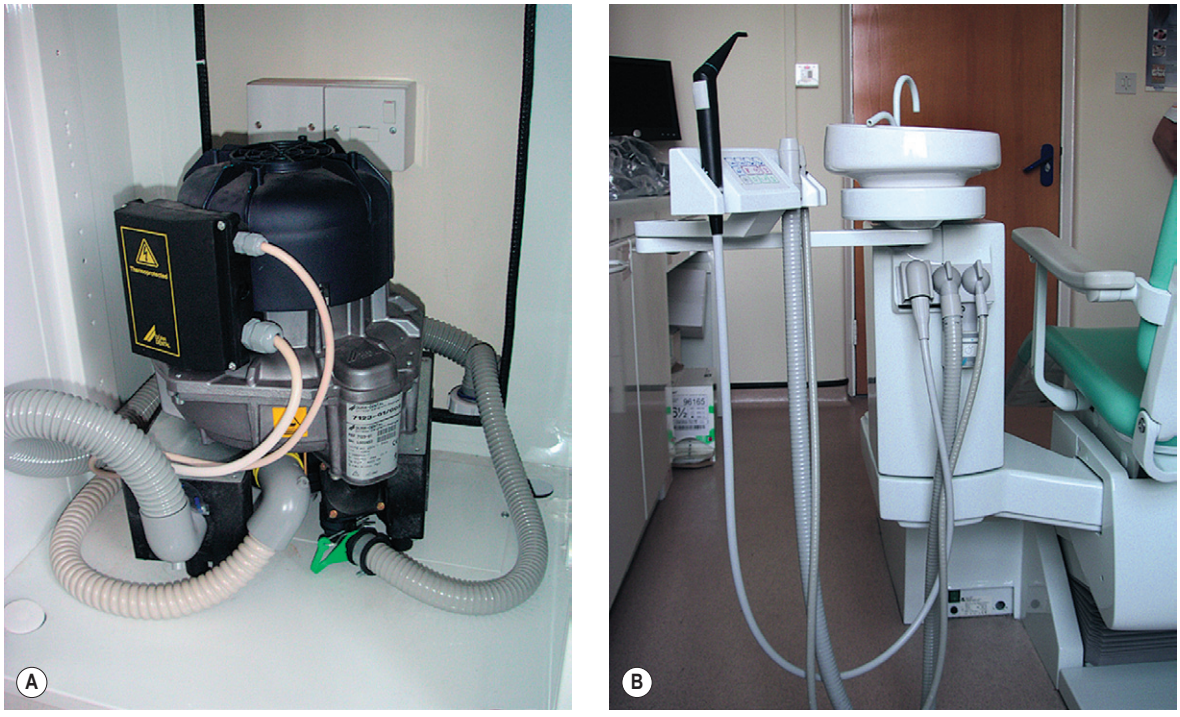


Figure 20.10 **A.** A vacuum pump in a dental surgery. **B.** A high-volume aspiration unit in a dental surgery. Note the different-sized suction tubes.

FURTHER READING

Medical Suction Equipment. Electrically powered suction equipment. Safety requirements. BS EN ISO 10079-1. London: British Standards Institute; 2009.

Medical Suction Equipment. Manually powered suction equipment. BS EN ISO 10079-2. London: British Standards Institute; 2009.

Medical Suction Equipment. Suction equipment powered from a vacuum or pressure source. BS EN ISO 10079-3. London: British Standards Institute; 2009.

HTM 02-01. *Medical Gas Pipeline Systems.* Department of Health Estates and Facilities Directorate. London: The Stationery Office; 2006.

Chapter 21

Cleaning, disinfection and sterilization

Trevor A King and Richard PD Cooke

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INTRODUCTION

All staff involved with anaesthetic equipment have a duty of care to ensure that the risk of hospital-acquired infection from anaesthetic equipment is kept to an absolute minimum. To this end in the UK, guidelines on infection control in anaesthesia were first published by the Association of Anaesthetists of Great Britain and Ireland (AAGBI) in 2002 and updated in 2008.[1] Also in the UK, healthcare organizations now have a legal responsibility to implement changes to reduce healthcare-associated infections (HCAIs). The Health Act 2006[2] provided the Healthcare Commission with statutory powers to enforce compliance with the Code of Practice for the Prevention and Control of Healthcare Associated Infection.

Although there are only a limited number of published reports relating to cross-infection,[3,4] the AAGBI recommendation was that all reusable anaesthetic equipment must be appropriately decontaminated prior to patient use and single-use items must be discarded immediately following use. Unfortunately, wide variation in decontamination practices between anaesthetic departments is well recognized. This can be avoided by ensuring that every hospital has a comprehensive infection control policy in place for all anaesthetic equipment, with a nominated anaesthetist and infection control doctor (ICD) taking lead responsibilities.[5] Such a policy should be evidenced-based and subject to periodic audit and review. Other useful UK web-based resources include the National Resource for Infection Control and the NHS Library (Surgery, Theatres and Anaesthesia Specialist Library). Key areas requiring risk assessment are listed in Table 21.1.

The aim of this chapter is not to be prescriptive, but to provide the necessary background information for a hospital to formulate its own anaesthetic equipment infection control policy in order to minimize the risk of HCAIs and comply with the UK's 'Health Act 2006'.[2]

RISK ASSESSMENT AND THE DECONTAMINATION PROCESS

Contaminated medical devices are typically classified into three infection risk categories:[6]

1. *High risk (critical) devices* – items in contact with a break in the skin, mucous membranes or introduced into a sterile body area. Such items must be sterile at the time of use. Examples include surgical instruments, dressings, catheters and prosthetic devices.
2. *Intermediate risk (semi-critical) devices* – defined as devices in contact with intact mucous membranes or contaminated with readily transmissible organisms (which do not penetrate skin or enter sterile parts of the body – see high risk devices). Disinfection is required though sterilization is preferred if the devices are heat-stable. Examples include endoscopes and respiratory equipment.
3. *Low risk (non-critical) devices* – items in contact with healthy intact skin. Cleaning, typically using hot water

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Table 21.1 Clinical areas where anaesthetic equipment is used

Theatres	Special care baby units
Day surgery units	Accident and emergency departments
Endoscopy suites	Radiology suites electroconvulsive
Maternity units (including neonatal resuscitaires)	Therapy suites
Coronary care units	Ward resuscitation stations
Intensive care and high dependency units	Ambulances
Respiratory units (non-invasive ventilation)	

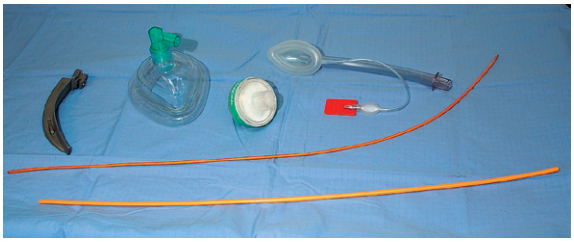


Figure 21.1 A selection of single-use anaesthetic equipment.

and a neutral detergent or a disposable detergent wipe, and drying are adequate for such items. Examples include non-disposable ECG electrodes, sphygmomanometer cuffs and stethoscopes.

If disposable anaesthetic equipment is chosen (Fig. 21.1), then a risk assessment may not be required. However, it is important to remember that disposable equipment will be for 'single use' or 'single patient use' only. 'Single use' indicates that the manufacturer intends the item to be used once only on an individual patient and then discarded. The packaging will be labelled either 'Single use', 'Do not re-use' or with the symbol ②.

'Single patient use' indicates that the manufacturer advises that the item may be used more than once on the same patient.[7] Examples of 'single patient use' items in anaesthesia include ventilator tubing and bacterial/viral filters used in critical care units.

By its nature, reusable anaesthetic equipment poses an intermediate or high (if used on broken mucous membranes or skin) risk of infection. Consequently, decontamination by sterilization or disinfection is required. However, the responsibility for choosing the correct decontamination method lies with the sterile services department (SSD) manager, supported by the ICD, the lead clinician and the relevant manufacturer's guidance. This is because the choice of decontamination method will depend on a number of factors, including the nature of the contamination, the time required for processing, the heat, pressure, moisture and chemical tolerance of the item, the availability of the processing equipment and the risks associated with the decontamination method. Furthermore, decontamination performed in SSDs will ensure that procedures are undertaken in a controlled and standardized manner, as well as being subject to audit review. Consequently, detailed knowledge of decontamination processes is not required by users. However, a sound understanding of the principles and level of decontamination required is essential, to ensure that a local infection control policy is both practical and supported by all staff involved in intensive care and anaesthetic practice. This also applies to the purchase or loan of any new anaesthetic equipment, since a preliminary decontamination assessment must be made prior to its use on patients.

Terminology

Decontamination

This refers to the removal or destruction of *contaminants* (*soiling with potentially infectious or other unwanted material*). Decontamination involves either cleaning alone, cleaning followed by high-level disinfection or cleaning followed by sterilization.

Bioburden

This is the population of viable infectious agents contaminating a medical device and should be routinely assessed by SSDs.

Cleaning

This is the physical removal of infectious agents or organic matter. It involves washing with a solvent (usually water and detergent), which may be heated (e.g. thermal washer disinfection, Fig. 21.2). This process does not necessarily destroy infectious agents. It is an essential process prior to disinfection or sterilization to remove bioburden. Items may also be placed in a water bath incorporating an ultrasound generator (Fig. 21.3). The ultrasound causes the water to vibrate at high frequencies so that it literally 'shakes' off organic matter. Ultrasonic washers are particularly useful for equipment where conventional cleaning methods might not reach some aspects of the device, or where items are too delicate to be physically scrubbed (e.g. some ophthalmic instruments).

Disinfection

This process reduces the number of viable infectious agents, but does not inactivate all microbial agents or bacterial spores. Low-level disinfection kills most vegetative bacteria (except mycobacteria and endospores), some fungi and some viruses. High-level disinfection kills vegetative bacteria (but not endospores), mycobacteria, fungi and viruses.

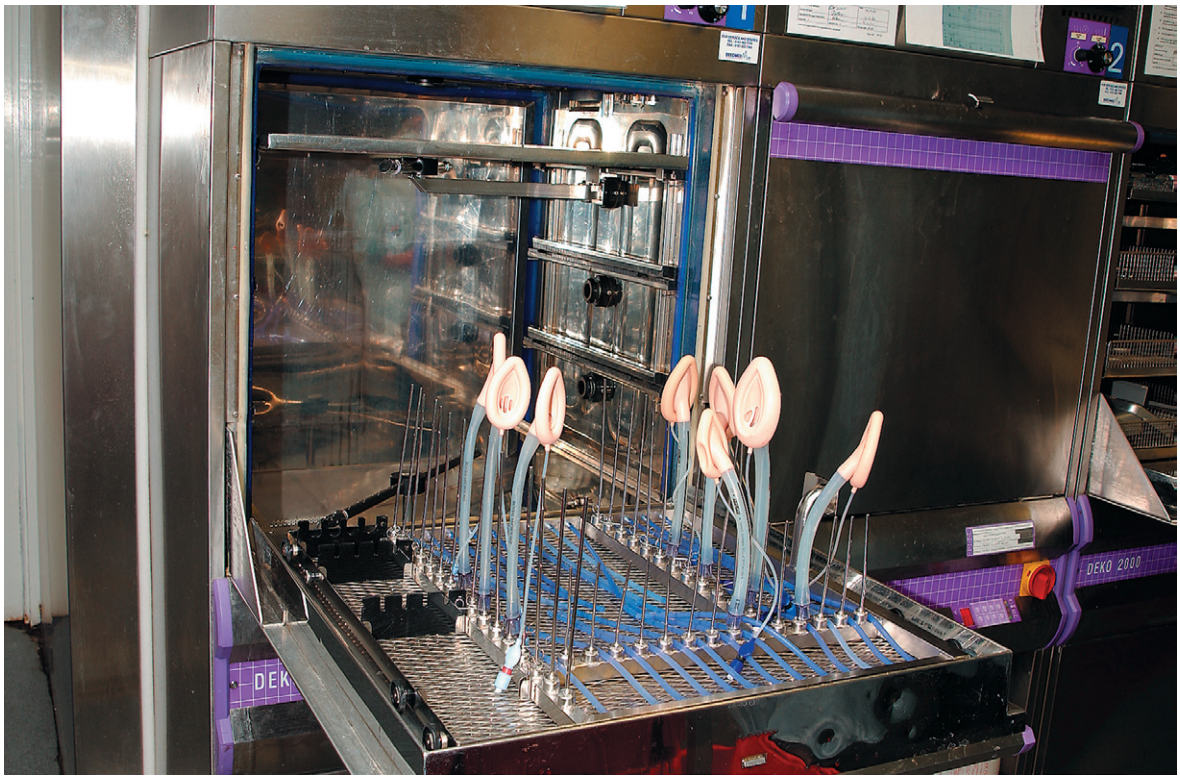


Figure 21.2 Laryngeal masks going into washer-disinfector.

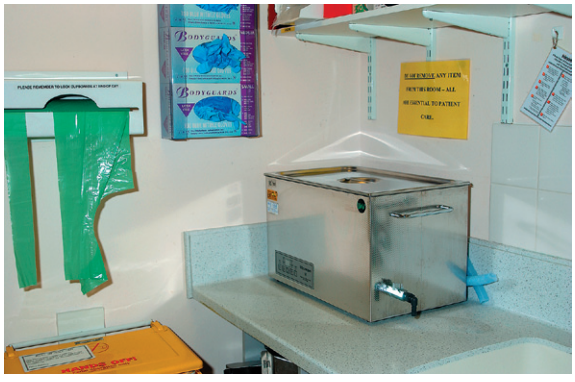


Figure 21.3 Ultrasonic washer used for endoscope accessories.

Sterilization

This is a validated process used to render a device free from infectious agents, including viruses and bacterial spores. Prions, which are recognized to cause transmissible spongiform encephalopathies (TSEs), are resistant to inactivation by conventional disinfection methods and sterilization procedures.

Sterilant

This is a liquid chemical agent, which can kill bacteria, fungi, viruses and bacterial spores (e.g. gluteraldehyde, chlorine dioxide, peracetic acid). However, this term is not precise and is not generally used. The term *high-level disinfectant* is preferred. A flow chart of appropriate decontamination methods is shown in Fig. 21.4.

Dry saturated steam

This involves saturated steam that contains no water particles in suspension (Fig. 21.5). See also Fig. 21.4.

INFECTION CONTROL STRATEGIES

Factors to be considered

Single-use versus reusable anaesthetic equipment

Infection control measures have been greatly simplified with the introduction of single-use anaesthetic equipment. Decontamination issues, already outlined with reusable equipment, will not be relevant. However, when preparing

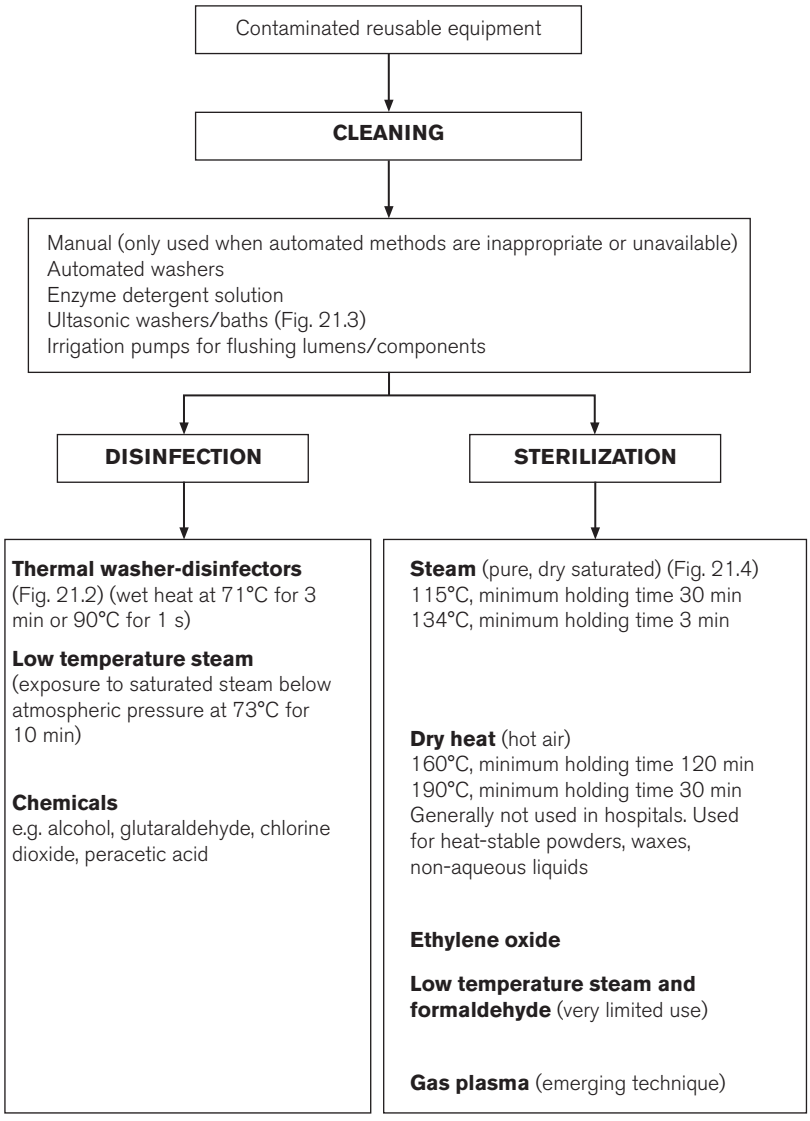


Figure 21.4 The decontamination process.[6]

an option appraisal, the revenue, procurement and waste disposal costs of single-use equipment needs to be balanced against capital procurement costs for new equipment and SSD charges for reprocessing items. In addition, careful thought needs to be given to where disposable equipment will be stored. To avoid compromising environmental cleaning standards, good communication is required between theatre and supply managers. An example of a local infection control policy for anaesthetic equipment is shown in Table 21.2.

Disinfection or sterilization

Decontamination by heat rather than by chemicals is always preferred. Disinfection by washer-disinfectors (Fig. 21.2) or low temperature steam is commonly employed, since autoclaving (with dry saturated steam) may damage anaesthetic equipment that is repeatedly processed.[5] Equipment used for invasive procedures or that comes into contact with broken skin or mucous membranes always requires sterilization.



Figure 21.5 Autoclave in sterile services department.

Centralization of decontamination services

When preparing an infection control action plan, it should be agreed at the outset with all relevant parties, that reprocessing of contaminated anaesthetic equipment should, whenever possible, be undertaken outside the clinical environment in central decontamination units or SSDs (Fig. 21.5).[1,6] Stand-alone autoclaves in theatres are not acceptable. Their use necessitates instrument cleaning (the most important component of the decontamination process) within the theatre complex. Furthermore, the scrupulous quality control measures required for the safe use of autoclaves cannot be guaranteed outside non-specialist areas.

Heat-labile instruments

Items of equipment needing high-level disinfection, but unable to tolerate high temperatures, such as endoscopes, pose particular infection control problems. Their decontamination should be undertaken in designated endoscopy units with appropriately trained personnel.[7] The use of chemical disinfectants, because of their toxicity, needs to be strictly controlled and subject to risk assessments in accordance with the Control of Substances Hazardous to Health Regulations, COSHH.[9]

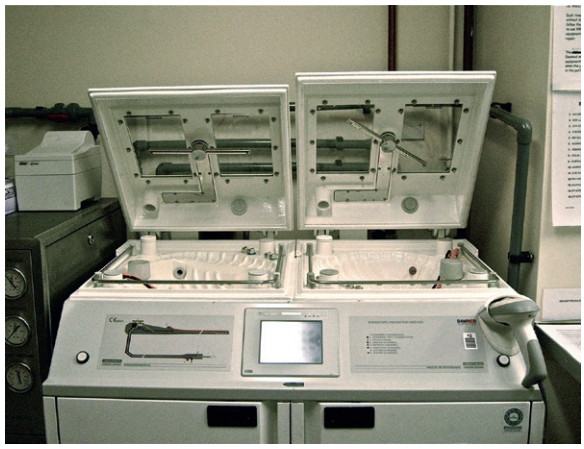


Figure 21.6 An automated endoscope decontamination unit. In the background (left) can be seen the associated water preprocessing unit.

Endoscope processing

Automated decontamination

Automated systems are expensive to purchase and maintain. They are produced by a number of manufacturers to reprocess heat-sensitive fibre-optic endoscopes to the satisfaction of the innumerable advisory and regulatory bodies. Intrinsic to most of these systems is the integration of some form of instrument tracking (see below). Individual manual processing involving gluteraldahyde baths and such like is no longer acceptable in UK hospitals.

Endoscopes first need manual cleaning before being attached into the machines. They are then processed at around 37°C through a cycle that starts with a leak test and continues through washing with enzymatic detergents, chemical disinfection and rinsing. Leak testing is necessary to ensure integrity of the internal lining and external wrapping of the fibrescope which would otherwise allow ingress of the caustic fluids and also pose an infection risk. The working channel(s) of the endoscope is connected into the washers so that it receives the same processing as the outside. Typical cycle times are of the order of 30 min (Fig. 21.6).

Drying cabinets

Concern in the UK, about contamination from airborne pathogens and the potential for growth of micro-organisms in stagnant fluid in endoscope channels, has resulted in recommendations requiring that endoscopes are processed within at most 3 h prior to use unless they have been stored under particular conditions. In practice this obligates the

Table 21.2 An example of a local infection control policy for anaesthetic equipment

EQUIPMENT	ACTION	COMMENT
Airways		
Oral/nasopharyngeal	Single-use	
Plastic endotracheal	Single-use	
Tracheostomy tubes	Single-use	
Angle pieces	Single-use	
Catheter mounts	Single-use	
Red rubber endotracheal	Return to SSD	
Reusable laryngeal masks	Return to SSD	40 uses maximum*
Single-use laryngeal masks (and other supraglottic airway devices)	Single-use	
Anaesthetic breathing systems (theatres)	Disposable	AAGBI recommends that anaesthetic circuits are changed on a weekly basis. A bacterial/viral filter should be used with every patient. Change between patients if visibly contaminated or used for highly infectious cases, e.g. tuberculosis.[4] External surfaces routinely cleaned with detergent wipes between cases.[13]
Circle absorbers	As for ventilators (see below)	Single-use bacterial/viral filter for each patient
Anaesthetic masks	Single-use	
Bougies		
Gum elastic	Return to SSD	Five uses maximum*
Others	Single-use	
Entonox delivery system (including mouthpiece/mask, tubing, one-way expiratory demand valve)	Single-use	
Laryngoscope blades and handles	Return to SSD or single use	
Manual resuscitators (self-inflating bags used at resuscitation)	Single-use	
Oxygen mask and tubing	Single-use	
Paediatric resuscitaire, facemask and tubing	Single-use	
Temperature probes	Single-use	
Ventilators (anaesthetic machine)	Routine disinfection not required except for external surfaces. Follow cleaning and maintenance policies for specific ventilators	In ICU, bacterial/viral filter placed on ventilator expiratory port and changed daily* or sooner since single-patient use only
Ventilator tubing (ICU)	Disposable	Change circuits weekly or sooner since single patient use only (see above)
Flexible fibre-optic laryngoscopes and bronchoscopes	Reusable following national guidelines[8]	Only reprocessed in a specialist endoscopy unit

*Manufacturers' recommendation; SSD, Sterile Services Department; ICU, Intensive Care Unit; AAGBI, Association of Anaesthetists of Great Britain and Ireland. (After King and Cooke 2001[3] with permission from The Hospital Infection Society.)



Figure 21.7 A typical HEPA filtered storage cabinet, here a Lancer Fibro-Dryer FD8.

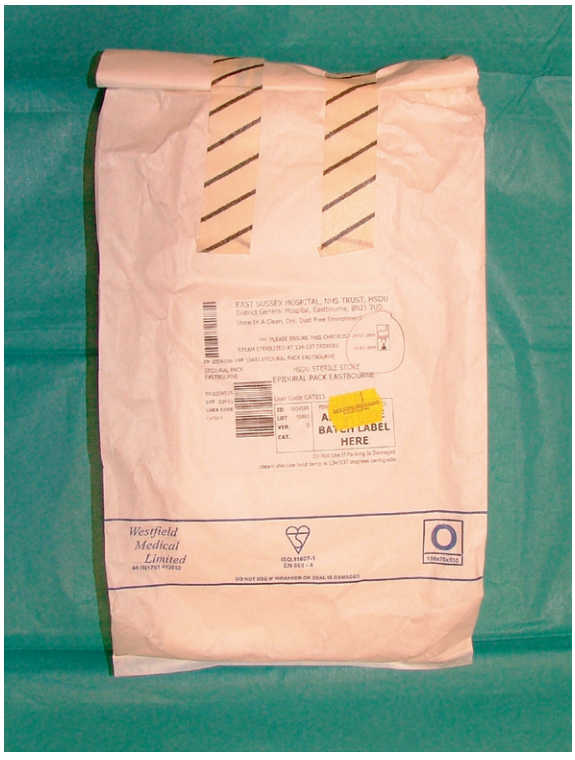


Figure 21.8 Sterile reusable epidural pack showing autoclave indicator tape and bar coded tracking labels. A second autoclave indicator is present inside the pack to confirm that adequate sterilization has taken place.

use of a dedicated automated drying cabinet, allowing storage times of 3 days to a week (Fig. 21.7).

High-efficiency particle absorbing (HEPA) filters are used to filter the air that is blown through the endoscope channels and the cabinets. HEPA filters, by definition, remove at least 99.97% of airborne particles 0.3 µm in diameter. Filters and cabinet are irradiated with ultraviolet light, which is bacteriostatic at adequate exposure levels.

Standard precautions

Universal infection control precautions should be used for the prevention of hospital-acquired infection between patients and staff. Classification of patients into high- and low-risk groups is inappropriate, since carriers of blood-borne viruses (BBVs) may be asymptomatic and unknown to staff. The same principles apply to the decontamination of medical equipment, the one exception being the prevention of prion-related diseases (see below).

The consistent application of effective hand hygiene is the cornerstone of good infection control practice and will minimize the risk of bacterial contamination of the operating theatre environment by anaesthetists (e.g. computer keyboard and telephones[10,11]).

Tracking of reusable anaesthetic equipment

SSDs must have tracking systems in place (Fig. 21.8) to enable reusable instruments to be traced to an individual patient in the event of a 'clinical incident'; for example, failure of the decontamination process or instruments being used on a patient with previously unsuspected Creutzfeldt–Jakob disease (CJD).[12] Furthermore, some reuseable anaesthetic equipment (e.g. laryngeal masks, gum elastic bougies) can only be reprocessed a finite number of times according to the manufacturers' instructions. A suitable tracking system therefore needs to be agreed between SSD and theatre staff, so that specific items of equipment are discarded after the maximum permitted number of uses. Single-use equipment overcomes the need for tracking systems to be in place.

Damage caused by the decontamination process

Some anaesthetic equipment, made from rubber or plastics, may be destroyed by autoclaving at 115–134 °C. Use

of washer-disinfectors operating at lower temperatures may then be the preferred option. Manufacturers will provide written instructions on the decontamination of reusable equipment and it is important that these are followed to avoid instrument damage. Some light sources for metal laryngoscope blades lose luminance with repeated high-temperature autoclaving.[14] This may influence the decision to procure single-use items.

Bacterial/viral filters and anaesthetic breathing systems

Bacterial/viral filters can be used to protect anaesthetic breathing systems from contamination. Although their role in the prevention of nosocomial pneumonia is controversial,[15] the AAGBI do recommend that a new bacterial/viral filter be used for every patient.[1] Similarly, implementation of a local policy on the reuse of breathing systems in line with manufacturers' instructions is recommended. Some breathing systems are marketed with instructions which permit reuse for a period of up to 1 week, if a new bacterial/viral filter is used with every patient. However, to ensure consistent application of infection control policies the AAGBI now recommends that anaesthetic circuits are routinely changed on a daily basis (see Table 21.2).

Prion disease

Abnormally shaped prion proteins are implicated as the causative agents in transmissible spongiform encephalopathies, of which variant (v) and sporadic CJD are examples. Both result in progressive neurological symptoms and death. Data from the CJD surveillance unit in Edinburgh indicates that deaths in the UK due to definite and possible cases of CJD amounted to 1476 from 1990 to March 2010.[16] Of these, only 168 are considered to be due to vCJD whilst sporadic CJD accounts for 1129 cases. Iatrogenic CJD (due to accidental transmission during medical or surgical procedures) has numbered between 0 and 6 cases per year and is mostly secondary to blood products and soft-tissue allografts. There have been no known transmissions of vCJD via surgery or use of tissues or solid organs.[17]

There remain a number of uncertainties concerning diagnosis, transmission and incubation periods for prion-related disease. However, the key infection control issue is the presence of microscopic traces of tissue which may remain on surgical instruments after standard decontamination procedures, which could transmit prion protein if inoculated into another patient. There has been no evidence of droplet-transmission. In both types of CJD, the highest concentration of prion protein occurs in the brain, spinal cord and posterior eye. With vCJD, the abnormal prion protein has also been detected in the appendix, tonsils, spleen and gastrointestinal lymph nodes.

The prion protein is remarkably resistant to conventional methods of disinfection and sterilization.[17] Standard washing techniques do reduce the concentration of prions in an exponential fashion, but 10–20 cycles are required to produce negligible levels. Standard decontamination methods will not protect against transmission of prion proteins, although the risk of transmission by surgical equipment from patients with undiagnosed CJD is thought to be minimal. Specific decontamination processes, which are the specialist realm of SSDs, have been recommended by the *Advisory Committee on Dangerous Pathogens and the Transmissible Spongiform Encephalopathy Working Group* and the National Institute for Health and Clinical Excellence.[12,17]

From the anaesthetic perspective, the transmission of prion disease can be reduced by the strict adherence to standard universal infection control precautions and the employment of single-use instruments when they are considered to be as reliable and safe as reuseable alternatives. To avoid contamination by tonsillar and adenoid tissue in adenotonsillectomy, laryngeal masks, bougies and other intubation aids should not be reused. However, the AAGBI does not recommend the mandatory use of disposable laryngoscopes if laryngoscopy could be difficult, in view of the extremely low risk of transmission.[1]

Equipment used on definite, probable, possible or 'at risk' CJD patients should be dealt with as follows:[17]

1. If used on patients with *definite or probable CJD*, the instruments should not be reused and must be destroyed by incineration or quarantined exclusively for reuse on the same patient.
2. If used on patients with *possible CJD*, reusable equipment should be quarantined until the diagnosis is either confirmed (when equipment must be destroyed) or excluded (following which equipment may re-enter general use). Suspect patients are defined as having clinical symptoms suggestive of CJD, but have not yet had the diagnosis confirmed.
3. For '*at risk*' patients, single-use equipment should be used wherever possible. Asymptomatic patients, who are potentially at risk of developing CJD or a related disorder include recipients of hormones derived from human pituitary glands, recipients of human dura mater grafts and those with family history of CJD. For procedures involving high- or medium-risk tissues (i.e. brain, spinal cord, anterior and posterior parts of the eye, lymphoid tissue, olfactory epithelium), instruments should be disposed of by incineration. However, for procedures on low-risk tissues, normal reprocessing guidance should be followed.

REFERENCES

1. Association of Anaesthetists of Great Britain and Ireland. Infection control in anaesthesia. *Anaesthesia* 2008;**63**:1027–36.
2. Department of Health. *The Health Act 2006: Code of practice for the prevention and control of healthcare associated infections. Revised 2008*. London: DoH Publications; 2008.
3 Neal TJ, Hughes CR, Rothburn MM, Shaw NJ. The neonatal laryngoscope as a potential source of cross-infection. *J Hosp Infect* 1995;**30**:315–7.
4. Mehtar S, Drubu YJ, Vijeratnam S, Meyet F. Cross-infection with *Streptococcus pneumoniae* through a resuscitaire. *Br Med J* 1986;**295**: 25–6.
5. King TA, Cooke RPD. Developing an infection control policy for anaesthetic equipment. *J Hosp Infect* 2001;**47**:257–61.
6. Medicines and Healthcare Products Regulatory Agency. *Sterilisation, disinfection and cleaning of medical equipment: guidance on decontamination from the Medical Advisory Committee to Department of Health. Parts 1, 2 and 3*. London: DoH; 2010.
7. Medical Devices Agency. *Single-use medical devices: implications and consequences of reuse. MDA DB 2000(04)*. London: Medical Devices Agency; 2000.
8. Medical Devices Agency. *Decontamination of endoscopes. MDA DB 2002 (05)*. London: Medical Devices Agency; 2002.
9. Health and Safety Executive. *Control of Substances Hazardous to Health Regulations 2002 (as amended, fifth edition): Approved Code of Practice and guidance*. London: HSE Books; 2005.
10. Jeske HC, Tiefenthaler W, Hohlrieder M, Hinterberger G, Benzer A. Bacterial contamination of anaesthetists' hands by personal mobile phone and fixed phone use in the operating theatre. *Anaesthesia* 2007;**62**:904–6.
11. Fukada T, Iwakiri H, Ozaki M. Anaesthetists' role in computer keyboard contamination in an operating room. *Journal of Hospital Infection* 2008;**70**: 148–53.
12. NHS Estates. *A guide to the decontamination of reusable surgical instruments*. Leeds: NHS Estates; 2003.
13. Baillie JK , Sultan P, Graveling E, Forrest C, Lafong C. Contamination of anaesthetic machines with pathogenic organisms. *Anaesthesia* 2007;**62**:1257–61.
14. Bucx MJL, Veldman DJ, Beenhakker MM, Koster R. The effect of steam sterilisation at 134°C on light intensity provided by fibrelight Macintosh laryngoscopes. *Anaesthesia* 1999; **54**:875–8.
15. Lorente L, Lecuona M, Málaga J, Revert C, Mora ML, Sierra A. Bacterial filters in respiratory circuits: an unnecessary cost? *Crit Care Med* 2003;**31**: 2126–30.
16. National Creutzfeldt-Jakob Disease Surveillance Unit. www.cjd.ed.ac.uk/figures.
17. Advisory Committee on Dangerous Pathogens and the Transmissible Spongiform Encephalopathy Working Group. Transmissible spongiform encephalopathy agents: safe working and the prevention of infection: part 4. London: DoH; 2010.

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Chapter 22

Information technology and the anaesthetic workstation

Chris J Barham

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Anaesthetists perform the majority of their clinical work in the operating theatre: the anaesthetic machine therefore, acts as their desk and office as well as a device for delivering anaesthesia. It is essential that it is equipped with the tools to provide care, record activity, provide information and enable communication.

This chapter will consider not only how information technology can assist in maintaining a record of the anaesthetic, but also its wider use in the theatre environment.

RECORD KEEPING

The earliest anaesthetic records date from 1894, although over 80 years later 3.4% of records in the UK were still merely an entry in the operating theatre register. It is now a legal requirement that an anaesthetic record is kept.

Most anaesthetists in the United Kingdom still create a handwritten record, which is sometimes augmented by a printout from the monitoring systems. These are both only stored in paper format. Table 22.1 lists the advantages and disadvantages of manual records compared with computerized records. The report of the National Confidential Enquiry for Perioperative Deaths for 2000[1] showed that 5% of case notes were lost, and in 3% of those present the anaesthetic record was missing. There is no evidence that these figures have changed significantly since then. Nevertheless, the report concluded that 'Improvements in Information Technology can make retrieval of patient information more, rather than less, difficult'. This must be viewed in the context of the time when records were all paper based, and the majority of information systems did not communicate with each other.

Functions of the anaesthetic record

In addition to the legal imperative for keeping an anaesthetic record there are many practical reasons for doing so.

Clinical communication

Most clinicians in the developed world work in a system that is based on corporate responsibility. For any patient episode in hospital, good clinical care often involves a number of individuals who are interdependent on one another's professional judgement.

Therefore, it is important that all information about the patient: including preassessment, preoperative evaluation, investigations, the intraoperative record and postoperative instructions and progress, is available to the appropriate clinicians.

Audit

Data from the anaesthetic record can provide valuable information about anaesthetic and surgical care. This can help to improve patient care and can assist in the management of the surgical process. Statistics can be provided for national and local use including information for the individual clinician for appraisal and review of performance.

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Table 22.1 Comparison of manual and computerized records

	MANUAL RECORDS	COMPUTERIZED RECORDS
Advantages	Familiar structure Easy navigation Portable Versatile and flexible Good 'user interface'	Available anywhere Available anytime Can be viewed by several users at the same time Accuracy of captured data On-line decision support Integration with medical record No need to enter data already present (e.g. medications) Ease of aggregate reporting
Disadvantages	Legibility Availability Accuracy Comprehensiveness Fragmentation No decision support Difficult to aggregate	Inflexibility Training required Update – new drugs, operations Data entry may be difficult Duplication of entries Distraction Different systems Artefacts

Education

Records can be used both as an educational tool, and to assess the adequacy of training. Trainees maintain log-books to demonstrate their experience, and this information can be better reviewed objectively and retrospectively from good-quality anaesthetic records.

Medicolegal

The record should be accurate, complete and legible, as its quality may be seen as a reflection of the quality of care given. It may also protect from litigation where the onus is on the practitioner to prove good medical practice.

Research

Extraction of information from clinical records is often essential in research.

Computerized anaesthetic records

Computerized anaesthetic record systems have been available for many years, but despite their benefits and the sophistication of modern systems they have yet to gain widespread acceptance. One of the major impediments to their introduction has been the lack of a 'business case' to prove their financial worth. This is despite the fact that anaesthetists are involved in 60% of inpatient hospital activity, and that many studies have shown the benefits of automated records.[2,3]

One of the difficulties in providing a business case is that there is little, if any, evidence of cash savings from the introduction of a computerized record. However, electronic patient records have been shown to provide many 'softer' benefits, including improved patient safety (due to reliable and clear communication), better information about the patient and the clinical process at the point of care, and the ability to identify good and poor care by linking care to outcomes.[4]

A modern computerised anaesthetic record system should also have comprehensive links to other clinical information systems to ensure that up-to-date information is available, and to avoid duplicate entry of pre-existing information (e.g. demographic details, proposed surgery, key personnel, etc.).

The National Health Service in England is currently making a major investment in information technology for integrated medical records through NHS Connecting for Health.[5] Initially anaesthetic and critical care systems were to be a key component, but delays in introducing the core systems have made it unlikely that anaesthetic systems will remain part of the programme. Now it is anticipated that the 'Clinical Five'[6] (see Box 22.1) will be delivered for most hospitals through off-the-shelf solutions, which

Box 22.1 The 'Clinical Five'

Patient Administration System (PAS)
Order communications and reports
Scheduling
Clinical correspondence with coding
Electronic prescribing

Box 22.2 Criteria for anaesthetic record systems

- Should link to:
  - Pre-assessment information
  - Electronic clinical record
  - Electronic prescribing system
  - Pathology
  - Radiology and PACS (public access computer system)
  - Theatre system for operating list information, personnel
- Perioperative record should have:
  - Appropriate user interface
  - Anaesthetic preoperative assessment record
  - Validation of staff
  - Capture of all patient monitor data
  - Capture of all machine monitor data
  - Configurable display of all trend data
  - Comprehensive data dictionary
  - Rapid entry of narrative from menus
  - Automatic coding
  - Free text entry
  - Drugs/fluids/infusions (including calculations)
  - Critical incidents
  - Postoperative instructions
  - Recovery progress
  - Key outcomes (death, pain, PONV, etc.)
  - Audit trail
- Reporting should include:
  - Staff logbooks
  - Activity analysis
  - Performance indicators
  - Data for audit
  - Financial analysis.

will be linked to each other and to existing systems (including anaesthesia and critical care) by some form of integration system.

Another of the key features of the NHS programme is the summary care record, which will carry summary information from the patient's medical record to be accessible at any site. This has now been developed and is being rolled out throughout England so the anaesthetist will have online access to key features of the patient's medical history, including medications and allergies. This should be available at any point of care including the anaesthetic work station.

The features required of a computerized anaesthetic system are listed in Box 22.2.

Automatic data capture

All displayed data from patient and machine monitoring systems (including infusion pumps, ventilators, etc.) should be captured by the system (Fig. 22.1). (Where target controlled infusion pumps are in use, data capture should include details of the pharmacokinetic model used together with the patient variables and the calculated drug concentrations.)

Automatic data logging is perhaps one of the principal benefits of computerised systems for anaesthetists, who may otherwise spend 10–15% of their time during a case on record keeping.[7] Some of this time may be saved and, more importantly, transcription errors are avoided. It may be argued that the act of keeping a manual record focuses the attention, but this is unsupported and is outweighed by the provision of a clear detailed graphical record (Fig. 22.2). Any anaesthetist keeping a manual record will be aware of the tendency during long cases for the interval between recordings to increase as the case progresses. An automatic record will maintain recordings with the same granularity throughout – including times when the anaesthetist is occupied directly with the patient.

The software for a computerized record system may be either a local application on the anaesthetic machine or a web-based application on a remote server. The latter is now the most common scenario for electronic patient record systems, generally as it is far easier to maintain. However, this does make data capture more complex as it must be relayed via the network. Specialist products are available to handle this.

The most robust configuration for a real-time continuous record system, such as that for anaesthesia, is a locally based application where the record is constantly mirrored to a central server. This ensures that it can continue in the event of a network failure, and in the event of a local terminal failure the record can be restored as soon as the terminal is replaced. However, the improved reliability of most networks and the ease of maintenance of client/server applications will make web-based solutions more common in the future.

Data entry

There should be an appropriate method of entering information to the system. This will normally be a keyboard or touch screen, together with some pointing device. These must all be suitable for use in the theatre, and should be easy to clean to avoid cross infection. Washable, sealed, plastic-coated keyboards, which may even be cycled through a dishwasher, are now available (Fig. 22.3).

Data entered should conform to the standards recommended by the Royal College of Anaesthetists (see Box 22.3),[8] and the system should be capable of attributing all procedures to the individual member of staff. Data



Figure 22.1 An established networked anaesthetic record keeping system in place on the workstation.

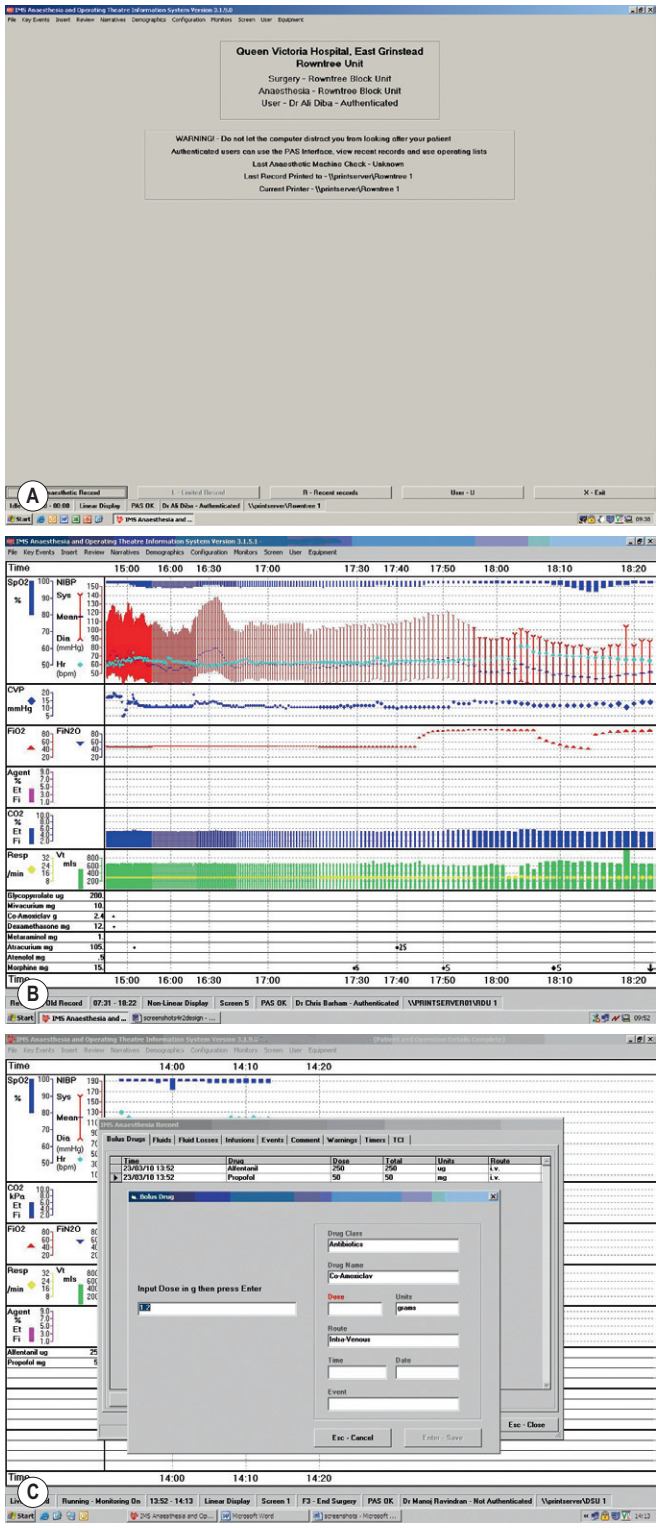


Figure 22.2 Screen shots from the type of record keeping system seen in Fig. 24.1: **A.** the splash screen; **B.** the trend page in use; **C.** a dialogue box for drug entry.

Box 22.3 **Suggested anaesthetic record set**

Preoperative information

Patient identity

Name / ID No. / Gender
Date of birth

Assessment and risk factors

Date of assessment
Assessor, where assessed
Weight (kg), height (m)
Basic vital signs (BP, HR)
Medication incl. contraceptive drugs
Allergies
Addiction (alcohol, tobacco, drugs)
Previous GAs, family history
Potential airway problems
Prostheses, teeth, crowns
Investigations
Cardiorespiratory fitness
Other problems
ASA +/– comment

Urgency

Scheduled – listed on a routine list
Urgent – resuscitated, not on a routine list
Emergency – not fully resuscitated

Checks

Nil by mouth
Consent
Premedication, type and effect

Peroperative information

Place and time

Place
Date, start and end times

Personnel

All anaesthetists named
Qualified assistant present
Duty consultant informed
Operating surgeon

Operation planned/performed

Apparatus

Check performed, anaesthetic room, theatre

Vital signs recording/charting

Monitors used and vital signs (specify)

Drugs and fluids

Dose, concentrations, volume
Cannulation
Injection site(s), time and route
Warmer used
Blood loss, urine output

Airway and breathing system

Route, system used
Ventilation: type and mode
Airway type, size, cuff, shape
Special procedures, humidifier, filter
Throat pack
Difficulty

Regional anaesthesia

Consent
Block performed
Entry site
Needle used, aid to nerve location
Catheter y/n

Patient position and attachments

Thrombosis prophylaxis
Temperature control
Limb positions

Postoperative instructions

Drugs, fluids and doses
Analgesic techniques
Special airway instructions, incl. oxygen
Monitoring

Untoward events

Abnormalities
Critical Incidents
Preop., perop., postop.
Context, cause, effect

Hazard flags

Warnings for future care

(Royal College of Anaesthetists Newsletter 36 (1997) – reproduced with permission.)



Figure 22.3 Washable keyboard incorporating a 'trackpad', suitable for clinical use. Man & Machine Computer Accessories Ltd, UK.

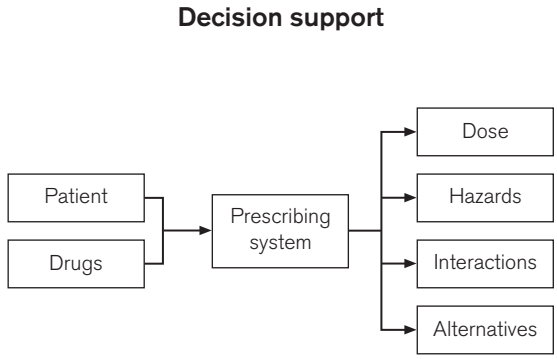


Figure 22.4 The process of decision support from a prescribing system based on embedded knowledge.

should also adhere to a standard schema and terminology to ensure information is comparable wherever it is collected.[9]

OTHER INFORMATION AND COMMUNICATION SYSTEMS

Electronic prescribing

If the hospital uses an electronic prescribing system, drugs recorded during anaesthesia should also appear in the main medication administration record. This is particularly important in avoiding the possibility of duplicate administration (e.g. antibiotics) or overdose (e.g. analgesics).

Decision support

Reference has been made above to the ability of electronic record systems to incorporate decision support. This can be defined as any method that takes input information about a clinical situation and then produces inferences that can assist practitioners in their decision-making. For example a prescribing system (and, hence, also an anaesthetic system) should be able to give the clinician information about dosage, interactions, and alternatives on the basis of embedded knowledge about the patient and drug (Fig. 22.4).

Access to local guidance (e.g. drug formularies and other policies) and information on the wider intranet (e.g. NHS evidence and Map of Medicine) should be available.

Pharmacology display systems

Decision support systems mentioned above should not be merely to warn of rare untoward drug interactions, but can also be made integral to the routine control of anaesthesia. Response surface pharmacodynamic interaction models can be used to guide anaesthetic drug dosing. By automatically acquiring data from infusion pumps, the anaesthesia machine and patient information systems, and by using inbuilt pharmacokinetic and pharmacodynamic models such '*advisory display systems*' can be made to predict pharmacodynamic responses and hence used to guide dosing of hypnotic, analgesic, relaxant and ultimately other drugs. Such a system is the SmartPilot View from Dräger (Fig. 22.5).

Communication

Until recently the only method of communication for the anaesthetist in the UK was a message via a third party, or possibly through an intercom system with, for example, the recovery ward. The use of mobile telephones in the operating theatre was, until recently, prohibited due to the perceived risk of electromagnetic interference causing malfunction of therapeutic and monitoring systems. Thankfully this is no longer the case as it is recognized that the benefits of mobile phones far outweigh the risks. It is vital that the anaesthetist can communicate with pre- and postoperative areas and many organizations invest in sophisticated communication systems using cordless phones or personal communicators that allow hands-free communication. In addition the appropriate use of email through the anaesthetic workstation can be of benefit in dealing with patient care.



Figure 22.5 Dräger SmartPilot View. See text.

REFERENCES

1. National Confidential Enquiry into Perioperative Deaths. *Then and Now. The 2000 Report of the National Confidential Enquiry into Perioperative Deaths*. London: NCEPOD; 2000. p. 29.
2. Devitt JH, Rapanos T, Kurrek M, Cohen MM, Shaw M. The anaesthetic record: accuracy and completeness. *Can J Anaesth* 1999;**46**:122–8.
3. Byrne AJ, Sellen AJ, Jones JG. Errors in anaesthetic record charts as a measure of anaesthetic perfomance during simulated critical incidents. *Br J Anaesth* 1998;**80**:58–62.
4. http://www.ehr-impact.eu/downloads/documents/EHRI_final_report_2009.pdf.
5. Barham C. IT in the NHS – where are we now? *Royal College of Anaesthetists Bulletin* 2010;**59**:38–40.
6. Department of Health. Health Informatics Review Report. 10 July 2008.
7. Allard J, Dzwonczyk R, Yablok D, Block FE Jr, McDonald JS. Effect of automatic record keeping on vigilance and record keeping time. *Br J Anaesth* 1995;**74**:619–26.
8. Lack A. *New College Guideline for Anaesthetic Records. Newsletter 36*. London: Royal College of Anaesthetists; 1997. p. 3.
9. Gardner M, Peachey T. A standard XML schema for computerised anaesthetic records. *Anaesthesia* 2002;**57**:1174–82.

Chapter 23

Electrical hazards and their prevention

Patrick T Magee

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MAINS ELECTRICITY SUPPLY

Most electromedical devices, including anaesthetic apparatus and monitors, are powered by mains electricity. It is, therefore, important for the anaesthetist to have an understanding of the principles of its provision and its hazards. For reasons of efficiency of power transmission, the mains provides an *alternating current* (AC) rather than a *direct current* (DC), which is normally provided by batteries. In DC, current flows steadily in one direction, while in AC, it flows rapidly back and forth at a frequency of 50 Hz in the UK, Europe and elsewhere in the world, and at 60 Hz in the USA. These frequencies may be a good choice for power transmission, but they are more hazardous to the user than other frequencies, including DC (see later).

In cables carrying DC, one cable is designated positive and the other negative, which is not the case with AC cables. Fig. 23.1A shows how a three phase, 16 kV primary winding of a substation transformer steps the voltage down to a three phase 240 V root-mean-square supply (325 V peak). It also shows how these secondary 240 V windings are linked together and connected to earth at the 'star point' (Fig. 23.1B). For each 240 V supply, therefore, one end is deemed 'live' and the end connected to earth at the star point is deemed 'neutral'.

Because of this earthing of the neutral conductor at the power station, any person or object who is also connected to earth would complete an electric circuit by touching the live conductor, even if no contact were made with the neutral one. Fig. 23.2 shows how, under certain conditions, the circuit connecting a patient to a live lead may be completed by, for example, an earthed diathermy plate, resulting in fatal electrocution. However, in most modern diathermy machines, this plate is isolated from earth as far as mains current is concerned (see Chapter 24). Furthermore, here in this example of a faulty monitor, an additional interruption of the neutral cable would result in the apparatus not working. However, because the live cable is still functioning, any contact with the ('live') casing would lead to electrocution of an inadvertently earthed user.

Stringent precautions should be taken to ensure that the polarities are correctly defined and connected for all mains electrical apparatus. Electrical accidents can be minimized by careful, regular maintenance by qualified personnel. It cannot be overemphasized that, if a fault exists, the apparatus should be removed from use and the services of a competent technician sought. The current international standard regulating electromedical equipment, IEC 60601-1, lays down quite specific testing regimens for electromedical equipment before use.[1,2]

The inclusion of a lead which connects the metal chassis, frame and enclosure of the apparatus to earth ensures that under faulty conditions the enclosure is prevented from becoming live, and is thus called the 'earth' lead. The faults in Fig. 23.2 show a break in the earth lead to the metal enclosure of the monitor (1); this allows a second fault within the monitor (2) to render the apparatus dangerous to the patient. This is discussed further in the section below, on Class I equipment.

Apparatus can be rendered safer by the inclusion of a fuse in the electrical circuit. This may be installed in the mains supply circuit, in the plug of the electrical lead to the apparatus, or in the apparatus itself. It usually consists of a fine

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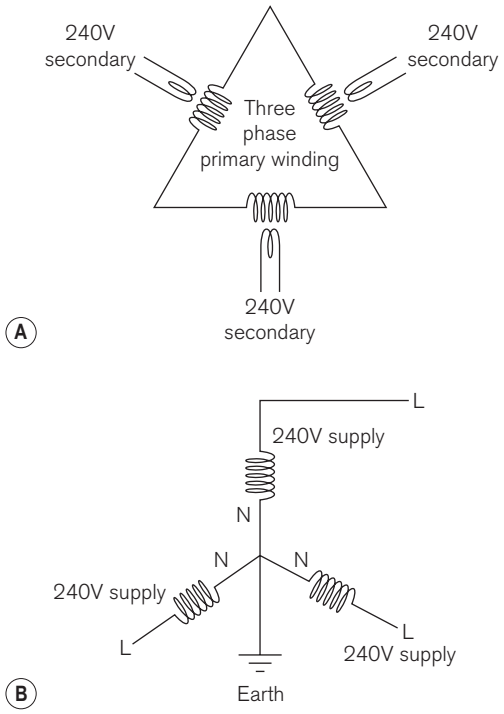


Figure 23.1 **A.** Power station transformer reduction of three phase 16 kV supply to 240V supply. **B.** 240V supply circuit at power station with neutral end of supplies connected to earth. L, live; N, neutral.

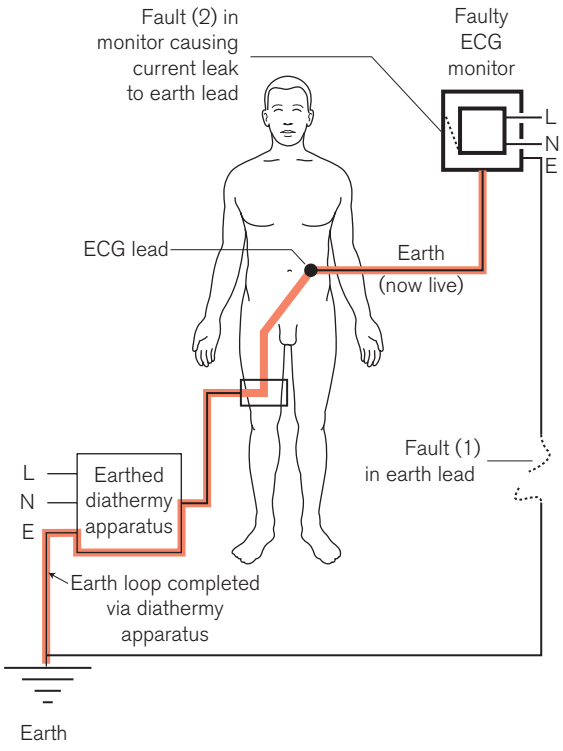


Figure 23.2 A fault in a monitor can cause electrocution by completion of a circuit (orange line) through another electromedical device.

gauge wire, which melts if the current passing through exceeds that against which they are intended to offer protection. So, in Fig. 23.2, if the earth lead of the monitor were intact, the fault consisting of a current leak between the apparatus and its enclosure, assuming adequate leakage current and low enough fuse rating, would result in a fuse in the live wire melting, breaking the continuity of the electrical circuit and the apparatus being rendered harmless. However, there is a risk that the fuse may not protect against electric shock. This can happen if someone is in contact with the equipment as the fault develops and before the fuse has time to melt (see below). Fuses are used mainly to interrupt the electric supply in the event that the current passing through the equipment exceeds a predetermined level that might cause overheating or damage. Other types of safety devices are mentioned later in the chapter.

PATHOPHYSIOLOGICAL EFFECTS OF ELECTRICITY

The pathophysiological effects of electric current passing through the human body include:

- resistive heating of the tissue and thermal burns
- electrical stimulation of excitable tissues, such as respiratory muscles and the heart
- electrochemical effects (electrolysis), when DC can cause chemical burns
- ignition of flammable material in contact with the body.

The factors that determine the severity of electrical injury are tissue resistance (R), current (I), potential difference (V), current frequency, current pathway and duration and current density. The thermal energy delivered to the tissues depends on the power dissipated (P), which can be calculated from:

$$P = V \times I = I^2 R$$

The body may be considered electrically to be an electrolyte (a good conductor) in a leathery bag (a poor conductor, an insulator). However, the resistance of the skin is very variable (Table 23.1).

Other tissues have diverse electrical resistances, which can be grouped as follows:

- *Low:* nerve, blood, mucous membrane, muscle
- *Intermediate:* dry skin
- *High:* tendon, fat, bone.

Table 23.1 Electrical resistance of skin

SKIN TYPE	ELECTRICAL RESISTANCE KΩ cm^{-2}
Mucous membranes	0.1
Vascular areas (volar aspect of arm, inner thigh)	0.3–10
Wet skin: in the bath	1.2–1.5
Sweat	2.5
Dry skin	10.0–40
Sole of the foot	100–200
Heavily calloused palm	1000–2000

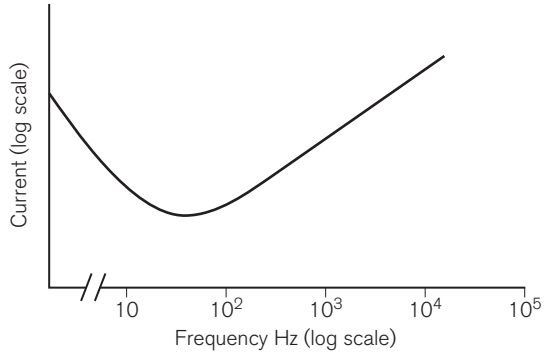


Figure 23.3 Current threshold variation with frequency for pathophysiological effects.

There is, however, an idiosyncratic relationship between whole body electrical impedance (AC resistance) and the applied voltage. At low voltages, 25–100 volts, it depends on the state of the skin and area of contact. At 250 volts and higher, the total body impedance falls to 2000–5000 ohm, irrespective of the contact area and the current pathway.[3]

The effects of electric current upon excitable tissues such as muscle and nerve depend not only on current and time, but also on the frequency.[4,5] It is one of the ironies of life that the commonly used mains frequencies of 50 Hz (UK, Europe) or 60 Hz (USA) are the frequencies at which the excitable tissues are at greatest risk of excitation and damage (Fig. 23.3).

In greatest danger is the heart, as it is susceptible to induced arrhythmias as well as permanent damage. The direction of the current pathway through the heart is also important. Clinical studies suggest that sudden death from ventricular fibrillation is more likely with current passing 'horizontally' from hand to hand, whereas heart muscle damage is more often associated with a 'vertical' current pathway.[6]

Table 23.2 The pathophysiological effects of 50 Hz AC current

CURRENT MA	PATHOPHYSIOLOGICAL EFFECT
1	Tingling
5	Pain
15	Severe pain and muscle contraction
30	'Let go' threshold
50	Respiratory muscle contraction, asphyxia
70	Multifocal beats, cardiac failure
100	Local burns, ventricular fibrillation
1000	Extensive burns, charring

The effects of hand-to-hand 50 Hz AC on the body are shown in Table 23.2 and Fig. 23.4. Fig. 23.5 shows a plot of current magnitude against duration in relation to pathophysiological effects.

Direct current (DC) electric shock tends to result in:

- single muscle spasm
- the victim being thrown from the source
- blunt mechanical trauma
- disturbance of heart rhythm.

Even very low imperceptible DC may produce electrochemical burns if the current is allowed to pass for long enough, for example from swallowed button-sized 1.5 V hearing aid type batteries.[7]

Alternating current (AC) electric shock:

- is about three times more dangerous than DC, at similar current flows
- produces continuous muscle contractions (tetany) at 40–110 Hz
- induces grip and pull as flexor muscles are much stronger than extensor muscles. If a person were to be holding onto a faulty conductor, he would be unable to let go. This prolongs the duration of the effect of the current
- induces local sweating, which reduces skin resistance.

ACCIDENTS ASSOCIATED WITH THE MAINS ELECTRICITY SUPPLY

As indicated above, there are four ways in which the mains electric current, or equipment powered by it, endanger the patient. These are:

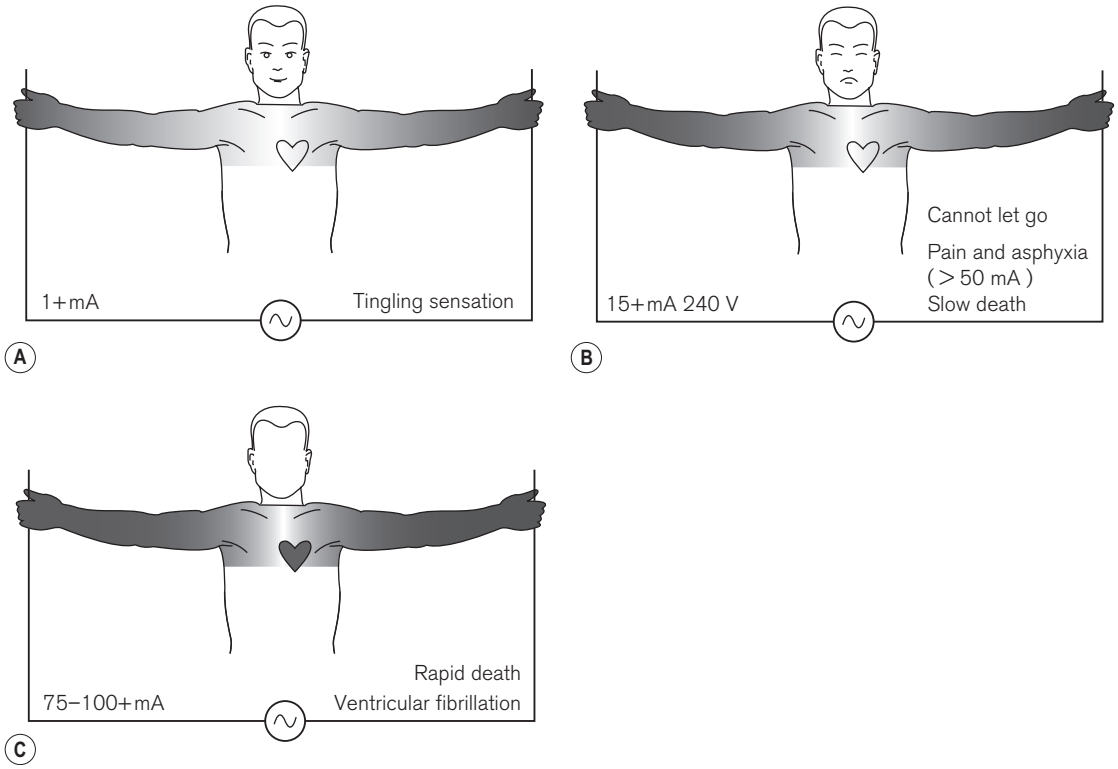


Figure 23.4 **A.** A current in excess of 1 mA passing through the body may produce a tingling sensation. **B.** If the current exceeds about 15 mA, muscles are held in tonic spasm, the victim cannot let go and will eventually die of asphyxia. **C.** When the current exceeds 100 mA, ventricular fibrillation and rapid death will occur.

- electrocution
- burns
- electrochemical effects
- ignition of flammable materials.

These will now be discussed in some detail.

Electrocution

As shown in Fig. 23.4, electrocution can cause death relatively slowly by tonic contraction of the respiratory muscles, leading to asphyxia, or more rapidly by ventricular fibrillation. The onset of ventricular fibrillation may be delayed, being preceded by ventricular tachycardia, which causes circulatory failure, but which may revert to normal rhythm if stopped in time.

As discussed earlier, the neutral pole of the mains electricity supply is connected to earth at the star point, a point at the power station which is thus remote from the patient. Since all conductors have some resistance, however low, there is therefore a small voltage drop between the patient end of the neutral conductor and the star point, i.e. they represent non-identical earth points. The patient end of the neutral conductor may, therefore, not be exactly at earth potential. This difference in potential along the neutral lead may facilitate stray capacitative or inductive currents in a circuit, which includes the patient connected to earth. Similarly, earthed electrodes may be attached to more than one part of the patient and from more than one piece of equipment supplied by different mains sockets, which may also facilitate stray capacitative or inductive currents in a circuit, which includes the patient. This is shown in Fig. 23.6. Therefore, it is recommended that the earth connections on all the socket outlets in a single clinical area be interconnected by a low-resistance conductor to minimize voltage differences between them. Similarly, all exposed metal objects, such as metal pipes and radiators, should be interconnected to a good earth.

Microshock

So far only *macroshock* has been discussed. Fig. 23.4 shows the effect of a current passing between the extremities. When it passes across the patient's trunk, only a small part of it passes through the heart. However, many modern medical, surgical and critical care procedures involve the

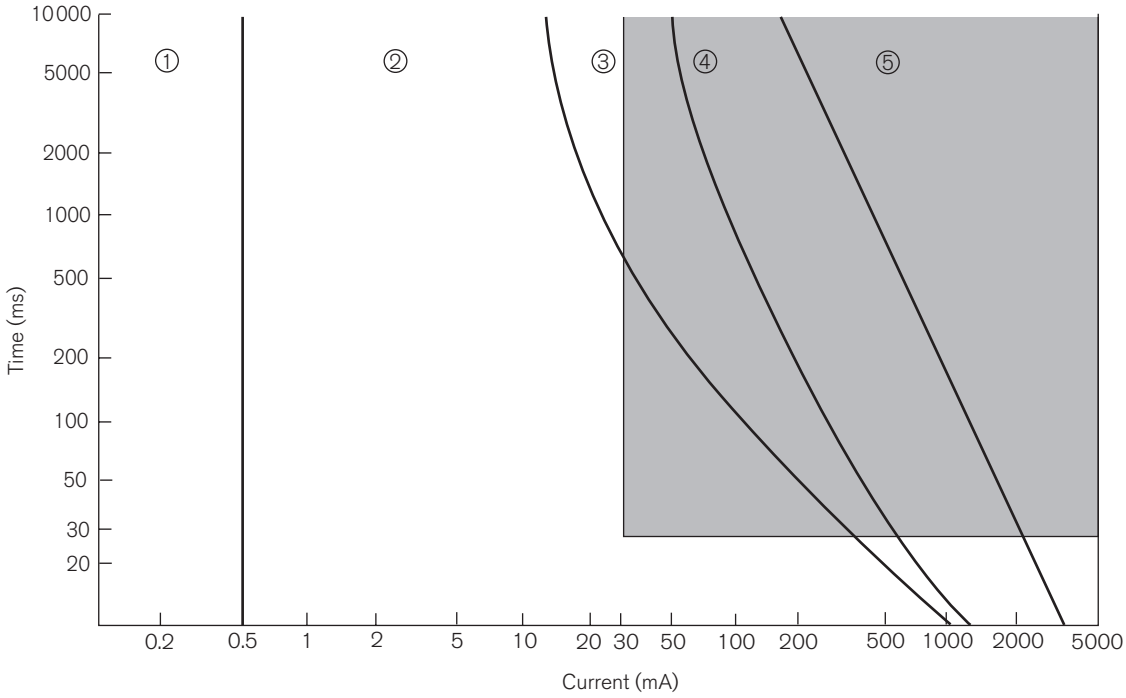


Figure 23.5 The effects of a current passing through the human body (hand to hand or hand to foot). Zone 1, usually no effect; zone 2, usually no dangerous effect; zone 3, usually no danger of ventricular fibrillation; zone 4, ventricular fibrillation possible; zone 5, ventricular fibrillation probable. The shaded area denotes the protection given by a current-operated earth-leakage circuit breaker (COELCB).

placement of electrodes on, within or close to the heart, e.g. a pulmonary artery catheter or even a transduced arterial line by virtue of its column of electrolyte. Under these circumstances, a very much smaller current, possibly as low as 100 μA, can result in ventricular fibrillation (Fig. 23.7) because all the current passes through the heart. A very small potential, such as the stray voltage in the mains neutral lead, could be sufficient to produce electrocution in this way. This phenomenon is known as *microshock*.

Shock protection

Apart from careful equipment design and construction and good equipment maintenance, as laid out in IEC standard 60601-1, there are two additional ways of preventing accidents caused by unwanted currents returning to earth:

1. Installing an isolating transformer, the output of which is carefully isolated from earth
2. Detecting unwanted currents passing to earth by a device that will sound a warning or automatically switch the supply.

Both have advantages and disadvantages. An isolating transformer may supply all the outlets for a whole

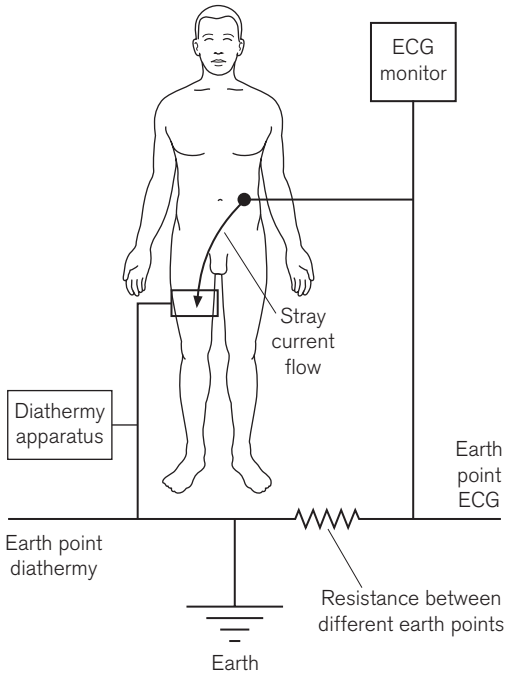


Figure 23.6 Stray current flow in a patient induced by earthing devices at different points.

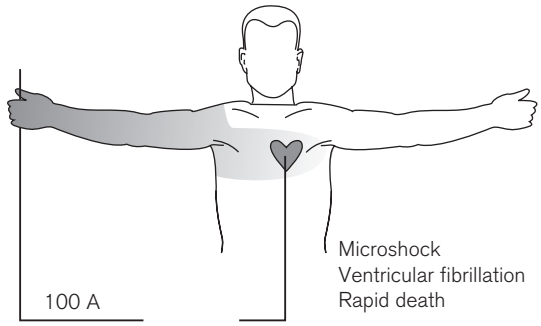


Figure 23.7 If one electrode is applied to the right ventricle of the heart itself, a very small current can result in ventricular fibrillation.

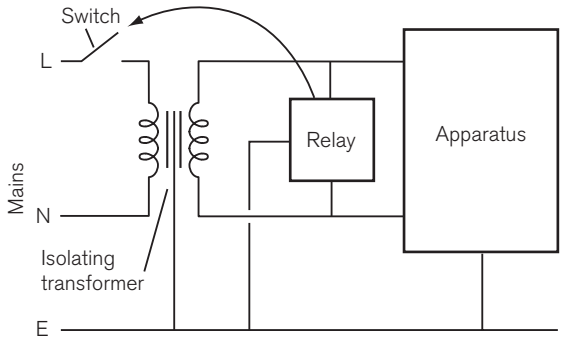


Figure 23.8 Safe patient power. The output of the isolating transformer is free from earth. Should earth leakage occur above a pre-arranged level, the relay will either disconnect the supply to the input of the transformer or sound a warning device. L, live; N, neutral; E, earth connectors.

operating room or theatre suite. It works on the principle that the output from the transformer is free from earth. Should the apparatus develop a fault, the earth leakage current is sensed almost instantaneously by a relay, which then trips a switch in the transformer input and cuts the power supply to it (Fig. 23.8). Apart from the expense, problems arise if there are several appliances in use and each of these has a small earth leakage current that is harmless in itself. The sum of all these currents may be sufficient to trip the relay and cut off the power to a monitor or other mains powered anaesthetic equipment. Likewise, a fault in one piece of apparatus may cause the power to another be cut off. If the relay operates an alarm rather than a circuit breaker, it may be ignored by staff. A better alternative is to include a small isolating transformer in the circuitry of each individual item of mains operated electromedical equipment, which can be connected to the patient. The patient circuit is, therefore, earth-free and said to be *fully floating*. The enclosure of the equipment may be earthed (or completely insulated see below).

The second method of improving safety is to install a *current-operated earth-leakage circuit breaker* (COELCB), also known as an *'earth trip'* or *residual current circuit breaker* (RCCB) (Fig. 23.9). This may be installed in the electrical supply to the whole operating room or theatre suite, or may be installed in each item of equipment. The live and neutral conductors each take a couple of turns or so (but both exactly the same number) around the core of a toroidal transformer. A third winding is connected directly to the coil of the relay that operates the circuit breaker. If the current in the live and neutral conductors is the same, the magnetic fields cancel themselves out. If they differ, there is a resultant magnetic field, which induces a current in the third winding and this causes the relay to operate and break the circuit. A difference of as little as 30 mA can trip the COELCB in as little as 30 ms or be used to operate an alarm. It may be manually reset and may also have a test button to check its operation. COELCBs may have similar problems to isolating transformers, but are less expensive. They operate so quickly and at such low earth leakage currents, that they greatly reduce the possibility of serious electric shock. The shaded area of Fig. 23.5 shows the protection afforded by a COELCB.

In the UK, electrical safety in clinical areas is achieved by a high standard of earthing of the fixed wiring, by good earthing of enclosures and fully floating patient circuits where appropriate. Safety may be further improved by using battery-operated equipment. In some cases the battery may be recharged from the mains between uses.

Classification of electromedical equipment to ensure electrical safety[1,8]

The international standard governing electromedical equipment, IEC 60601-1, is based on a concept of risk management; it tries to assess and control risk in the device design, manufacture and intended use. The standard requires that there be two levels of protection for the patient or, indeed, the operator of the equipment, so that even if one level fails, harm may yet be avoided. It allows the use of three mechanisms to provide those two levels;[9] these are insulation, protective earthing and protective impedance, and the following classifications use these mechanisms to different extents.

Class I equipment

This includes household electrical items. The electrical apparatus itself is connected to live and neutral conductors and insulated from the metal casing. The casing, which can be touched by the user, is connected to the earth lead,

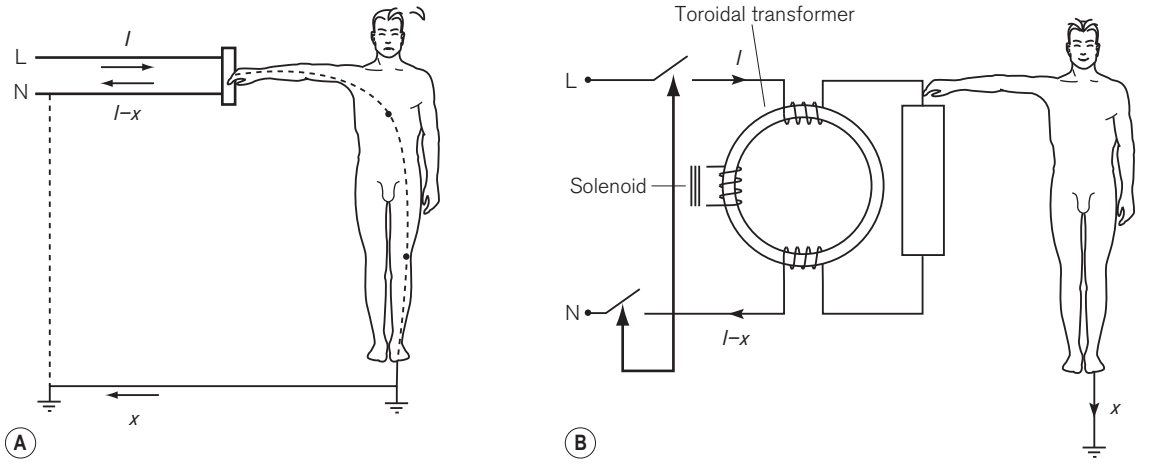


Figure 23.9 **A.** If a load is taking a current of *I* amps from the live conductor L and *x* amps is returning via the patient and earth, then the current in the neutral conductor N will be (*I* – *x*) amps. **B.** A current-operated earth-leakage circuit breaker (COELCB). The imbalance between the currents in the live L and neutral N conductors is sufficient to set up a field in the toroidal transformer sufficient to induce in the third winding, a current that will trip the solenoid and, therefore, disconnect both the live and neutral supply.

which normally safely conducts any *earth leakage current* to earth. In the event of a fault in the apparatus in which there is current leakage from the apparatus to its casing, the earth connection grounds this 'earth leakage current'. A fuse in the live wire plug connection (UK) melts and the apparatus becomes disconnected from the mains. The user is thus protected from electric shock. However, if the earth connection is also faulty, such protection is lost and there is a potentially hazardous *enclosure leakage current (touch current)* through inadvertent contact with the apparatus, or *patient leakage current* through the patient applied parts of the device[2] (Fig. 23.2). Testing of these various different leakage currents of the device before use is mandatory according to IEC60601-1. Not all devices with an earth lead are necessarily Class I; for example, some devices have an earth lead designed for screening the equipment, rather than carrying a leakage current. Furthermore, a device with an enclosure that appears to be made of plastic may still be Class I and should be tested as such.

Class II equipment (Fig. 23.10A)

Both the apparatus and its casing are insulated. The insulation may be a reinforced single layer or a double layer. There is no need to earth such equipment, but it should contain a fuse.

Class III equipment

This equipment is designed to operate from a power source with a voltage known as *safety extra low voltage* (SELV), defined as not exceeding 24 V AC, or 60 V DC. The apparatus may either have its own internal power source or be connected to the mains by an adaptor containing a transformer. Although macroshock is unlikely with such equipment, microshock is still possible. If battery-operated equipment can also be mains operated, for example for battery charging, or is capable of being operated via a mains powered transformer, then it must be tested as Class I or II equipment. Low voltage per se is not one of the mechanisms defined in IEC 60601-1 for ensuring equipment safety, and the standard does not explicitly refer to Class III equipment. Equipment that is not capable of being mains connected is referred to in the standard as 'internally powered'.

Class I, II, and III equipment may be subdivided into the following:

Type B equipment (Fig. 23.10B)

This defines a device in which the earth leakage current is restricted to 0.05 mA in Class I equipment or 0.1 mA in Class II equipment. It may refer to Class I, II or III, that is mains powered or internally powered. Although the equipment can be connected to the patient externally or internally, such equipment shall not be connected to the heart.

Type BF equipment (Fig. 23.10C)

This is as type B equipment, except that the part applied to the patient is isolated from the rest of the apparatus. This confers a degree of safety such that the maximum earth leakage current under a single fault condition is not exceeded, even if 1.1 times the maximum rated voltage is

the resistance of the above items remains within prescribed limits.

If flammable anaesthetic agents are used, the most important precaution, however, is the use of antistatic (conducting) rubber or neoprene in the components of the breathing systems and other tubing components in the anaesthetic machines. As recently as 1982, an explosion occurred when a co-axial breathing system, part of which was made of a non-conducting material, was used with cyclopropane.

Fire and explosion

For these to occur, there are three prerequisites: combustible material (fuel), oxidant to support combustion and a source of ignition.[10-12] These risks arise from the following sources:

- The use of high partial pressure of oxygen (pressurized oxygen and high oxygen concentrations)
- The use of flammable anaesthetic agents and of solvents for cleaning and skin preparation, such as alcohol.

When a fire starts, energy from the chemical combination of combustible material and oxygen is liberated in the form of heat. If it takes place in a confined space, the pressure from the hot gasses produced may increase greatly. Rapid liberation of heat and the rise of pressure result in an explosion.

High oxygen partial pressure

When there is a rise in the pressure of a gas, heat is generated. If a flammable material, such as oil or grease, in a confined area is suddenly subjected to a high partial pressure of oxygen, such as 14 000 kPa in a full oxygen cylinder, the heat generated is sufficient to ignite it and cause an explosion. (This is the principle of the compression-ignition (diesel) engine.) Therefore, oil, grease or other flammable materials should be kept away from apparatus in which high oxygen partial pressures exist.

Under some conditions, nitrous oxide may dissociate into nitrogen and oxygen and the latter gives rise to risk of explosion. Hence nitrous oxide sources should be treated with similar care.

Volatile anaesthetic agents and flammable solvents

Cyclopropane, most ethers and ethyl chloride are explosive in anaesthetic concentrations. They are no longer in common use in the developed world.

Carbon dioxide, halothane, enflurane, isoflurane, sevoflurane and desflurane are not flammable, nor is nitrous oxide at atmospheric pressure. However, nitrous oxide does support combustion even more fiercely than oxygen. Trichloroethylene is also non-flammable under conditions in which it is used by an anaesthetist. Halothane is flammable in oxygen, but at much higher concentrations than are used clinically.

Fire or explosion may be caused by the ignition of gasses or vapours within the anaesthetic equipment or escaping from it, or of the vapour of a flammable substance that is used for skin cleaning, such as alcohol.

Although naked flames are seldom employed in the operating rooms nowadays, they may be encountered elsewhere. However, ignition of flammable mixtures may be caused by sparks from static electricity, from electric motors, faulty electrical apparatus or from diathermy apparatus, electrical cautery (as used in ENT surgery), or from the electric plug top being pulled out of the socket when the switch is turned on and the current is flowing.

Other causes of fire

There are two other modern causes of burns and fire in the operating room: these are fibre-optic light sources and surgical lasers. Powerful visible light sources are now commonplace in the operating room to provide illumination for endoscopic procedures. These sources, even at the patient end of a fibre-optic light cable, produce a concentrated amount of heat as well as light. The carelessly placed end of the cable may cause a burn directly or set fire to the drapes or other flammable material. This situation is even more dangerous if the drapes have a high concentration of oxygen or flammable material beneath them.

Zone of risk

This is a term used to denote the area in which explosive mixtures are deemed liable to exist during anaesthesia. Within the zone of risk, the following precautions are advised:

- There should be no naked flames
- All electric switches should be spark-proof and electric plugs should be 'captive' while the switch is turned on
- All parts, especially rubber tubing, etc., of anaesthetic apparatus should be constructed of conductive (antistatic) rubber or other material, and the operating room floor should be antistatic. Antistatic rubber, containing carbon, has sufficient conductivity to leach away static electricity, and yet sufficient resistance to prevent so fast a discharge that a spark occurs.

All trolleys, stools and other mobile equipment should have tyres or feet of a conducting material. These are painted yellow or have a yellow flash or label to indicate that they are antistatic.

In 1956 in UK the original working party looking into the risks of explosion in clinical settings defined the zone of risk as the whole anaesthetic room and operating room where the anaesthetic machine was mobile. Since 1956, with non-flammable anaesthetic agents largely superceding cyclopropane and ether, there was a dramatic fall in the number of explosions. Subsequently, in 1970, the Association of Anaesthetists of Great Britain and Ireland changed the definition of the zone of risk to 25 cm around the gas pathways of the anaesthetic machine and breathing system. Non-spark-proof switches and sockets are permissible outside the zone of risk, providing they are permanently attached to the wall of the operating room, and that electrical outlets are 40 cm above the floor to prevent damage to cables.

Classification of anaesthetic equipment and zone of risk (Fig. 23.11)

This section is relevant only to those areas of the world where the use of flammable anaesthetic agents still occurs.

Anaesthetic-proof equipment (AP)

AP equipment standards are based on the energy required to ignite the most flammable mixture of ether and air. AP equipment can be used within a 5–25 cm range of flammable gas escaping from a breathing system, and its temperature should not exceed 200°C.

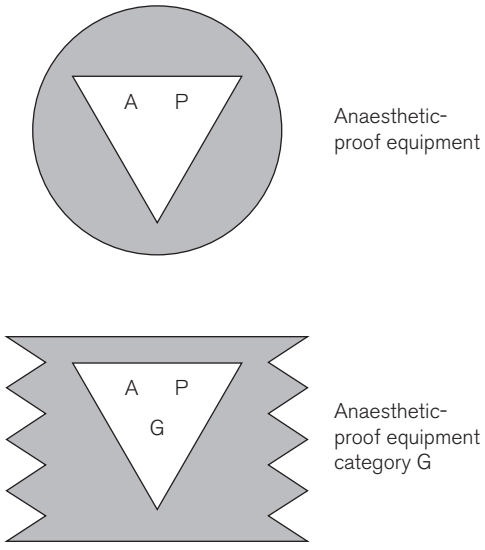


Figure 23.11 Symbols indicating 'zone of risk' classification for anaesthetic equipment.

Anaesthetic-proof category G (APG) equipment

These heightened standards are based on the ignition energy, which should not exceed 1 μJ, required to ignite the most flammable mixture of air and oxygen. APG equipment can be used within 5 cm of gas escaping from a breathing system, and its temperature should not exceed 90°C.

REFERENCES

1. IEC. *International standard. IEC 60601-1. Medical electrical equipment – Part 1: General requirements for basic safety and essential performance*. 3rd ed. Geneva: Bureau Central de la Commission Electrotechnique Internationale; 2005.
2. http://www.ebme.co.uk/arts/safety/part6.htm
3. Beiglemeyer G. Effects of current passing through the human body and the electrical impedance of the human body. In: *A guide to IEC Report 479*. Berlin: vde Verlag gmbh; 1987.
4. IEC. Effects of current passing through the human body. In: *IEC 479–1 General aspects*. Geneva: Bureau Central de la Commission Electrotechnique Internationale; 1984.
5. IEC. Effects of current passing through the human body. In: *IEC 479–2 Special aspects*. Geneva: Bureau Central de la Commission Electrotechnique Internationale; 1987.
6. Fontanarosa PB. Electric shock and lightening strike. *Ann Emerg Med* 1993;**22**:378–87.
7. Yoshikawa T, Asai S, Takekawa Y, Kida A, Ishikawa K. Experimental investigation of battery induced esophageal burn injury in rabbits. *Crit Care Med* 1972;**25**: 2039–44.
8. Al-Shaikh B, Stacey S. *Essentials of anaesthetic equipment*. 2nd ed. London: Churchill Livingstone; 2002. p. 175–80.
9. Eisner L, Brown RM, Modi D. *Safety requirements: understanding. IEC 60601-1*. Geneva: Medical Electronics Manufacturing Archives; 2004.
10. MacDonald A. A short history of fires and explosions caused by anaesthetic agents. *Br J Anaesth* 1994;**72**:710–22.
11. MacDonald A. A brief historical review of non-anaesthetic causes of fires and explosions in the operating room. *Br J Anaesth* 1994;**73**:843–6.
12. Vickers MD. Hazards in the operating theatre. Fires and explosions. *Ann R Coll Surg Engl* 1973;**52**:354–7.

FURTHER READING

Hull CJ. The electrical hazards of patient monitoring. In: *Monitoring in anaesthesia and intensive care*. London: Hutton & Prys Roberts Saunders; 1994. (Although now a little out of date, this chapter explains many aspects of electrical safety well.)

Magee P, Tooley M. Environmental and electrical safety. In: *The physics, clinical measurement and equipment of anaesthetic practice*. Oxford: Oxford University Press; 2005.

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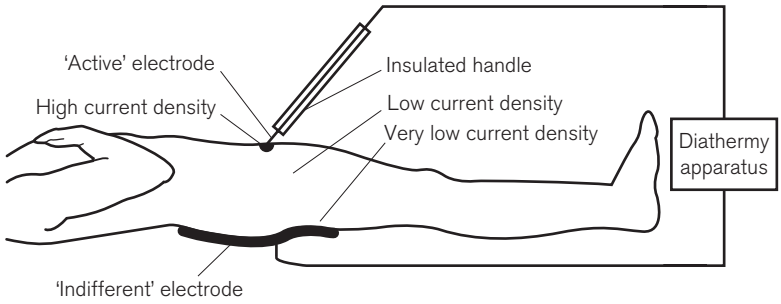


Figure 24.1 The concept of 'current density' or the current per unit cross-sectional area. In areas of low-current density the heat generated is quickly dissipated, whereas in the area of high-current density the heating effect is very high.

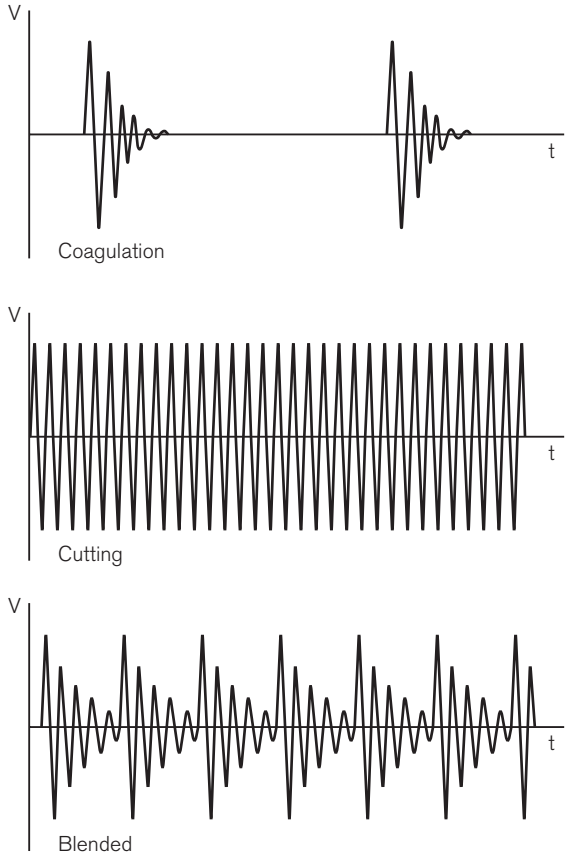


Figure 24.2 The waveforms commonly used for surgical diathermy.

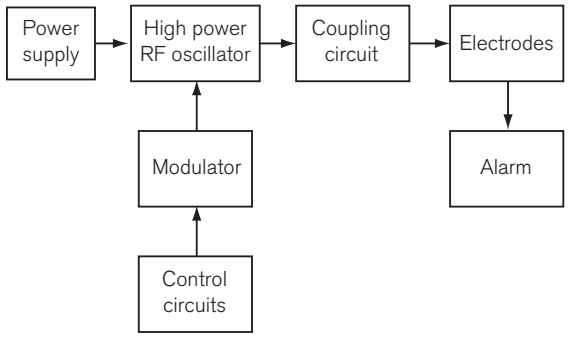


Figure 24.3 Block diagram of a surgical diathermy unit.

machine is shown in Fig. 24.3. A high-power, high-frequency oscillator or generator is controlled by a modulator to produce the necessary waveforms. The output of the generator is led through coupling circuits to optimize impedance matching between the generator and the 'indifferent' and the 'active' electrodes. The circuitry normally contains a 0.01 μF capacitor from the indifferent electrode to earth. It will be recalled that the electrical impedance (the term used to describe electrical resistance at different frequencies of current) is inversely proportional to both the frequency of the current and the magnitude of the capacitance; therefore, this capacitor in the diathermy circuit provides a low impedance (20 Ω) route to earth for the high-frequency diathermy current, but a high impedance (300 kΩ) to any incidental low-frequency mains current. This effectively isolates the circuit from earth against mains leakage current. The risk of electrocution from a diathermy apparatus is, therefore, much reduced, but the risk of burns should not be underestimated (see below). Where the indifferent lead (i.e. the plate) socket is at earth potential, as it is in many older diathermy sets, it is vitally important that the lead is connected to the correct terminal of the diathermy apparatus. If the plate were accidentally connected to the active terminal, when the foot switch is depressed the patient might be burned at all points connected to earth (i.e. where the body is in contact with those parts of the operating table at earth potential).

The above description is of a unipolar arrangement. However, some diathermy sets are capable of being used with a bipolar system in which the current passes from one blade of a pair of forceps to the other. The circuit is

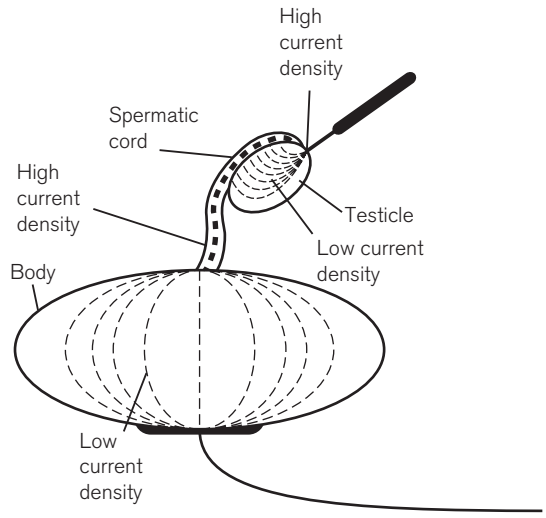


Figure 24.5 The testis elevated from the body on the spermatic cord generates unintentional areas of high current density (thick dashed lines) during unipolar diathermy.

One property of a capacitor is its decreasing impedance (the term used to describe frequency-dependent electrical resistance) and, therefore, increased current flow as the frequency of the AC current increases. This means that high-frequency diathermy currents are a significant potential hazard for burns in this respect. A capacitor may be formed between the diathermy probe and a surgeon's finger holding that probe, with the surgeon's glove acting as the intervening insulator.[5] Burns may be caused to the surgeon's gloved fingers in contact with the probe by capacitative coupling if the capacitatively induced current through the surgeon's finger finds a pathway to earth (for example through his boots, if there is not perfect insulation to the floor, or through his contact with another earthed surface). Another example of a capacitor is an insulated diathermy probe in contact with the vaginal wall during cervical cautery; burns to vaginal tissue might occur by capacitative coupling if the induced current finds an earth pathway, which it might do despite best attempts to avoid this; note that no assumption is made in these examples concerning an earthed, a non-earthed or a faulty diathermy circuit. Another example of possible capacitative coupling can occur during laparoscopic surgery (see below). The probe and the surrounding tissue act as capacitor plates, with the probe's insulation between them; capacitative coupling of high-frequency diathermy current may result in a tissue burn at this site rather than at the probe tip when the diathermy is activated.

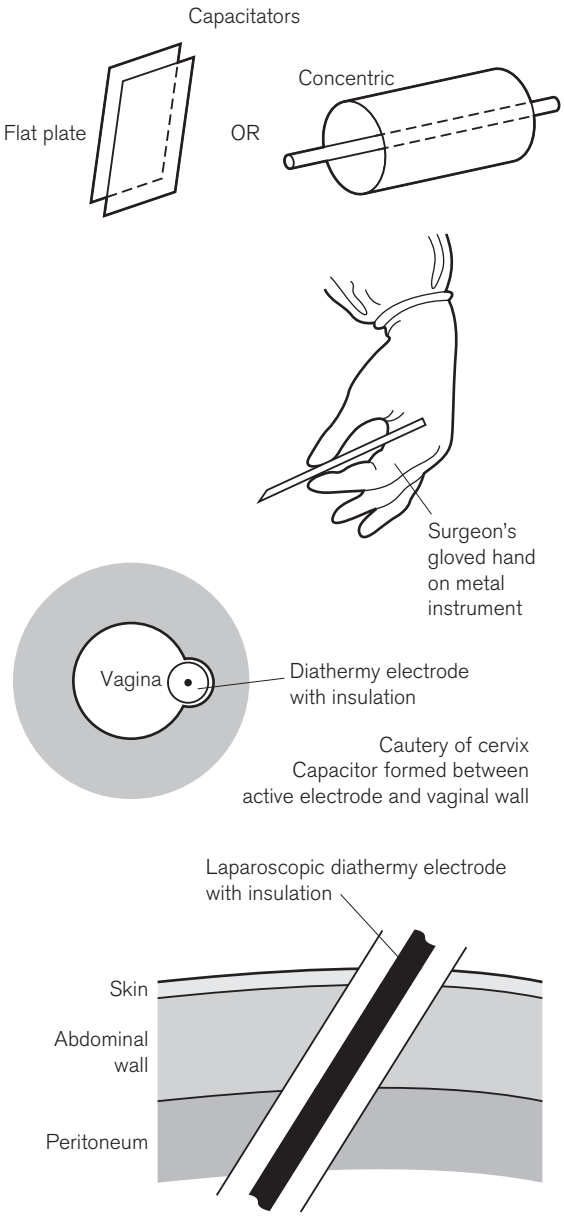


Figure 24.6 The concept of capacitative coupling. A capacitor is formed between any conductive surfaces separated by an insulator (dielectric). At DC, a capacitor is a pure insulator; as the frequency increases, its resistance (reactance, impedance) decreases. Inadvertent capacitors may be formed between an instrument and the surgeon's fingers, the active electrode and other tissue (e.g. the vaginal wall) or between the (insulated) active electrode and the abdominal wall during laparoscopic surgery.

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Chapter 25

Pacemakers and defibrillators

Nicholas P Gall

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SUMMARY

- Temporary pacing can be achieved in a number of ways; none are ideal.
- The indications for permanent pacemakers and implantable defibrillators are expanding. These devices are becoming increasingly complex.
- Strong sources of electromagnetic interference, diathermy for instance, can interact with pacemakers and defibrillators.
- Consultation with a cardiac electrophysiologist or cardiac technician is essential to understand the detailed functioning of these devices and to plan the treatment of patients.

BASIC CARDIAC ELECTROPHYSIOLOGY

While pacemakers and defibrillators are not strictly anaesthetic equipment, they are frequently encountered both in the elective and the emergency situation. An understanding of their hardware and software is important, not least because of their interaction with the strong electromagnetic fields that may be encountered in anaesthetic practice.

To understand pacemaker and defibrillator functioning, an understanding of normal cardiac electrophysiology is necessary. The myocardium consists of an interconnected network of myocytes. At rest, the myocyte interior is maintained at a negative potential (−80 mV) in relation to the extracellular fluid. This is due to the relative impermeability of the plasma membrane, differential intra- and extracellular ion concentrations and active ion transporters. Either because of the spontaneous inward leak of positively charged ions, in the sinus node for example, or because of an external electrical stimulus, the potential difference across the membrane decreases. At a threshold potential (approximately −70 mV in ventricular myocytes) various ion channels become activated producing further ion influx. This leads to myocyte depolarization and the *action potential* (Fig. 25.1). Calcium influx leads to the release of additional intracellular calcium stores activating the contractile apparatus, a process described as *excitation-contraction coupling*.

Further ion movements, particularly potassium efflux, restore the myocyte's potential to its resting value. During the early part of the action potential the myocyte cannot be induced to depolarize again – *the absolute refractory period*. Later on, a stimulus of sufficient strength can induce further depolarization – *the relative refractory period*.

Myocytes in different areas of the heart have different ion channels and, therefore, different action potentials. Some cells spontaneously discharge due to a resting inward leak of positively charged ions, with some cells, for example in the sinus node, discharging at a faster rate than others; this is known as *automaticity*. Due to the

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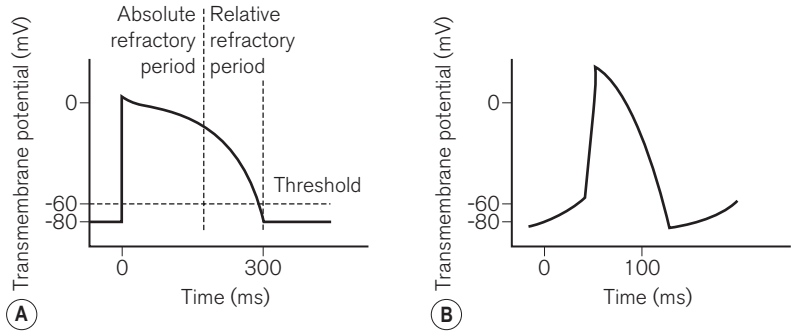


Figure 25.1 An action potential of a ventricular myocyte is shown on the left with that of a sinoatrial myocyte on the right.

inter-linked nature of myocytes, the cells with the fastest rate of depolarization set the heart rate.

PACEMAKERS

Introduction

The association between a slow pulse and syncope was recognized at least 300 years ago. With the development of the electrocardiogram at the turn of the twentieth century, the various forms of bradycardia were defined. At the same time experimental electrical stimulation of cardiac cells was first studied, followed soon after by the first descriptions of the use of cardiac electrical stimulation as a therapeutic modality. Dramatic improvements followed in battery, capacitor and electronic technology, driven particularly by the Second World War. On 8 October 1958 the first entirely implantable epicardial cardiac pacemaker was inserted. A transvenous device followed in 1962. Multiple technological advances have followed, leading to the complex and reliable pacemakers we have today.

Pacing essentially involves passing a small electric current into the myocardium, usually with a wire that is placed endocardially. The stimulus, if it is of sufficient strength and outside the absolute refractory period, can induce myocyte and, subsequently, chamber depolarization.

The NASPE/BPEG code

With the increasing complexity of pacemakers, a three letter code was designed in 1974 by the Inter-Society Commission for Heart Disease Resources to define a pacemaker's characteristics. This was refined in 1981 and subsequently led to the 1987 North American Society of Pacing and Electrophysiology (NASPE)/British Pacing and Electrophysiology Group (BPEG) five letter generic code.

The first letter of this code denotes the chamber(s) that are paced, the second letter the chamber(s) that are sensed, the third letter the response to sensing and the fourth letter programmability and rate response. The fifth letter, rarely used, denotes anti-tachycardia functions. NASPE/BPEG defibrillator (1993) and lead (1996) codes have also been designed.

In general terms, five forms of pacing are likely to be encountered clinically, the first four being available for both temporary and permanent pacing:

1. **VVI**. A single lead is placed in the ventricle, for pacing (**V**VI) and sensing (V**V**I). If the ventricular rate is above the minimum set (*the base rate*), pacing is *inhibited* (VV**I**). Otherwise pacing occurs.
2. **AAI**. A single lead is placed in the atrium. It functions in a similar way to the VVI and is used for sick sinus syndrome where AV nodal conduction is normal.
3. **DDD**. Leads are present in the atrium and ventricle. Both chambers (hence *dual*) can be paced (**D**DD) and sensed (D**D**D). If activity in either chamber is sensed a pacing stimulus will be inhibited. If no ventricular activity follows atrial activity, ventricular pacing will be *triggered* (DD**D** representing the dual responses of inhibition or triggering).
4. **VOO/AOO/DOO**. These are the simplest pacing modes. No sensing occurs (_**O**_) and therefore there can be no response to sensing (__**O**); the chamber(s) is paced at a fixed rate. This mode is used temporarily where the sensing of external noise could lead to the inappropriate inhibition of pacing, e.g. diathermy. It is also known as asynchronous pacing.
5. **VDD**. Only one lead is used which paces the ventricle (**V**DD). An electrode further up the lead is positioned in the atrium allowing dual chamber sensing (V**D**D). This allows physiological pacing, i.e. ventricular activity follows atrial at a physiological PR interval as long as atrial pacing is unnecessary. This mode is uncommon and is used as an alternative to DDD pacing in order to reduce system complexity.

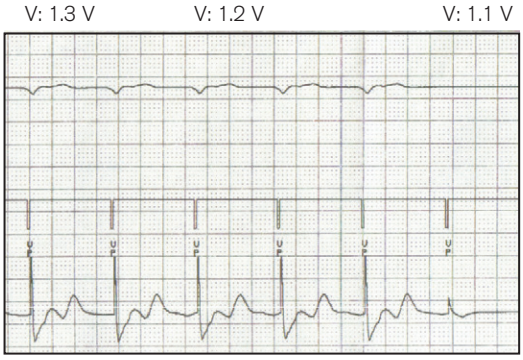


Figure 25.2 A printout from a VVI pacemaker is shown. A standard ECG lead is seen at the top. The pacemaker marker channel is seen beneath, VP indicating ventricular pacing. The bottom line shows the electrogram seen by the bipolar lead. Loss of capture is seen at 1.1V.

6. The presence of an (**R**) as the fourth letter denotes a pacemaker able to increase the paced heart rate in response to the patient's activity (see below, Software).

Pacing terminology

A number of pacing terms require explanation:

- *Threshold* (Fig. 25.2) – the minimum stimulus needed to capture the heart. It is given as current (amperes) or voltage (volts) over a time period (pulse width, ms).
- *Unipolar/bipolar* – refers to the arrangement (Fig. 25.3) of the negatively charged cathode and the positively charged anode. In the bipolar circuit, the cathode and the anode are separated by a short distance at the pacing lead tip. The circuit is localized to the immediate myocardium. In the unipolar circuit, the cathode is at the pacing lead tip with the anode at a distance; in the permanent pacing system it is found in the pacing box. The unipolar pacing circuit is larger and, therefore, produces a larger pacing artefact on the ECG. While there is little difference between the pacing characteristics of these two conformations, bipolar sensing is less susceptible to external noise.
- *Impedance* – this represents the total resistance to current flow in the pacing circuit. This resistance occurs in the leads, at the lead–myocardial interface for endocardial leads and in the tissue between the cathode and the anode. Fractures in the leads increase impedance, while insulation breaks reduce it.

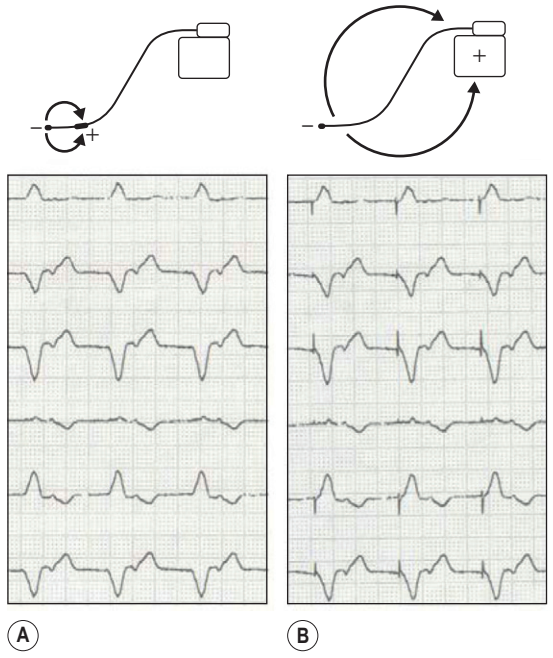


Figure 25.3 Bipolar (**A**) and unipolar (**B**) pace/sense circuits are shown. Representative ECGs are shown beneath; note the larger pacing artefact on the unipolar ECG due to the larger pacing circuit.

Temporary pacing

Temporary pacing is used in emergency situations of life-threatening bradycardia or where a bradyarrhythmia could occur temporarily.

Transvenous pacing

This is the usual method for temporary pacing. A thin, semi-flexible, shaped, bipolar pacing lead is normally used. The central circulation is accessed via the internal jugular, subclavian or femoral veins, the first of these being preferable as there is a lower risk of complication (e.g. pneumothorax with the subclavian route, infection and deep vein thrombosis with the femoral). Furthermore, with the internal jugular approach, the veins used for permanent pacing are not directly punctured. The lead is manipulated into position at the right ventricular apex under X-ray guidance. Some leads may have a central shapeable stylet to allow more accurate positioning. Others have a tip-mounted flotation balloon to allow non-radiographic positioning. Active fixation (screw tipped) leads are also available to prevent lead displacement. In most cases one ventricular lead is used. Atrial pacing and dual chamber pacing are also possible. For temporary transvenous pacing, due consideration must be given to the possible need for permanent pacemaker insertion and

the impact thereon of infection and local complications at vascular access. It is hence advisable to avoid the subclavian route and to use ultrasound guidance for vascular puncture. Various guidelines are listed in the bibliography section at the end of this chapter.

The pacing lead is connected to an external box which allows various programming options:

1. Mode: usually VVI ('demand') or VOO ('asynchronous' or 'fixed rate')
2. Output: either the output voltage or current is programmable, the pulse width being fixed. Maximum outputs are higher than with permanent systems as battery size is unrestricted and the threshold is more unstable
3. Sensitivity: the threshold above which electrical activity will be detected as cardiac. This is usually measured in millivolts, a higher value indicating lower sensitivity
4. Rate: usually 30–150 or even higher for the overdrive pacing of tachyarrhythmias.

Transoesophageal and transgastric pacing

An electrode (on a device similar to an oesophageal stethoscope or nasogastric tube),[1] placed in the oesophagus, can capture the left atrium, while in the stomach ventricular pacing may be possible. Transoesophageal atrial pacing (TAP) is a simple and safe method for temporary treatment of bradyarrhythmias and is particularly applicable to use during anaesthesia.[2] Because voltages of up to 20V or more may be required to achieve capture (Fig. 25.2), a signal amplifier is needed if using an ordinary external (transvenous) pacing box (signal generator). TAP, like transgastric pacing, is rarely used because of unfamiliarity with the technique and scarcity of equipment.

Transcutaneous pacing

In the peri-arrest situation it may be difficult to position a transvenous temporary wire. It is possible to capture the ventricle by passing a large enough current through the chest using specially designed electrodes (Fig. 25.4). These electrodes may also be used for monitoring and defibrillation.

There is significantly greater impedance in the transcutaneous pacing circuit than with endocardial stimulation, due to the additional significant impedances of the electrode–skin interface, the lung and the pericardium. High currents are required, usually 50–90 mA, at pulse widths of 10–20 ms. Pacing thresholds may be further increased by poor electrode–skin contact, metabolic disturbance (hypoxia, acidosis) and pericardial effusion. Unpleasant cutaneous nerve and skeletal muscle stimulation occur at currents as low as 10 mA making this method of pacing, without sedation, short-lived by necessity. Pulse widths of the order of 10 ms produce optimal pacing thresholds and reduce patient discomfort. Significant pacing artefacts occur on the ECG, making myocardial capture difficult to assess without pulse or blood pressure monitoring.

Another form of transcutaneous pacing is also used by cardiothoracic surgeons. One or more wires are implanted directly into the atrial and/or ventricular myocardium at the time of operation and brought out through the skin. Two wires can be used per chamber to allow bipolar pacing, or a second wire may be stitched into the praecordial skin as and when pacing is actually required. These can be connected to a temporary pacing box. Similar to temporary wires they are prone to infection and thresholds can rise without warning. They are usually used to await recovery of the normal conducting system. Atrial pacing may also reduce the incidence of postoperative atrial fibrillation.

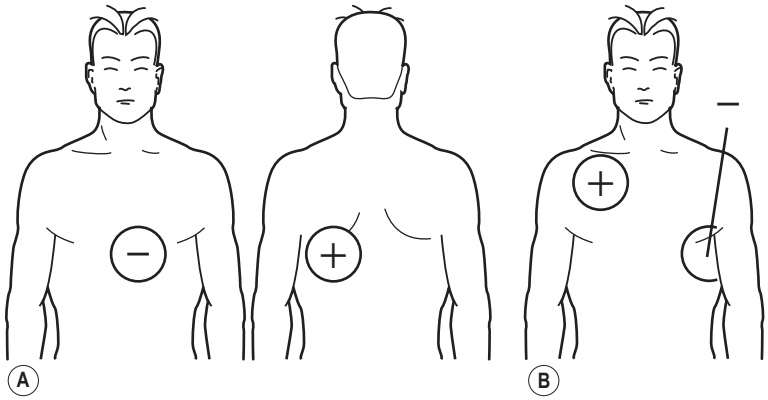


Figure 25.4 The suggested positioning for transcutaneous pacing electrodes is shown, antero-posterior (**A.**) and antero-lateral (**B.**). There is no significant difference in pacing thresholds between the two conformations.

Permanent pacing

The principles of permanent pacing remain the same as temporary pacing; the indications continue to expand and have been recently updated. Following significant improvements in battery and lead technology, devices are now small enough to be placed subcutaneously in the pre-pectoral region, the leads passed via the subclavian or cephalic veins. Other routes including the femoral veins and epicardial systems may also be used where subclavian access is impossible or due to previous pacemaker infection.

Hardware

The pacemaker box consists of:

1. The battery. This forms a major part of the volume of the pacemaker. It has a number of important requirements including:
   - **a.** small size
   - **b.** the ability to produce small amounts of current reliably and consistently over long periods
   - **c.** a low self-discharge rate so that it does not wear out without use
   - **d.** a predictable discharge rate so that replacement can be planned in advance
   - **e.** lack of gas production to allow hermetic sealing
   - **f.** significant longevity to allow infrequent box changes. It is not unusual for batteries to require replacement only once a decade.

A number of different power sources have been used, including nuclear devices. However, the lithium-iodine battery is currently the industry standard.

2. Voltage/output circuits. Various circuits are required to allow the battery voltage (usually 2.8 V at implant) to be altered to produce a range of programmable outputs, according to the threshold, to prolong battery life.
3. Telemetry circuits. These allow an external computer to communicate and programme the pacemaker using radiofrequency signals.
4. Pacing circuits. These control all aspects of pacemaker function including the timing circuits, diagnostics and the rate response circuitry.
5. Sensing circuits. These allow sensing of intrinsic cardiac activity with filtering allowing exclusion of external electromagnetic noise.
6. Memory. Modern pacemakers are able to store significant quantities of data, for example on arrhythmias and heart rate variability.
7. Reed switch. This is activated by the magnet, completing a circuit allowing a particular predefined programme to start.

In contrast to temporary leads, permanent pacing leads are expensive, highly engineered multi-component devices, required to be *in situ* for several decades, without failing or producing significant local damage. Their components include:

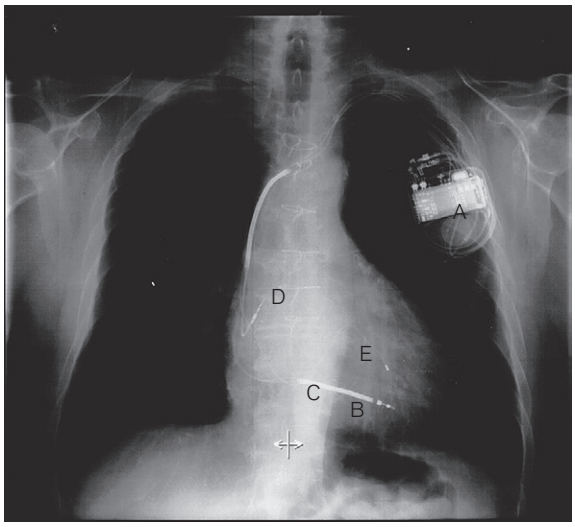


Figure 25.5 A chest radiograph of a biventricular ICD is shown. A number of the components are visible: the box (A), the screw tip of the active-fixation ventricular lead (B), the distal shocking coil (C), the atrial lead (D) and the coronary sinus left ventricular pacing lead (E).

1. Electrode. Most newly implanted leads are bipolar with a cathodal tip and a larger anode further back. The cathode is surprisingly complex – it is small, producing: a high charge-density, thus improving myocardial capture; and a high pacing impedance which reduces battery drain. Platinum, titanium or activated carbon is used for their good conducting properties and durability. A small reservoir of steroid is also found in the lead tip to reduce the inflammatory reaction at the lead–myocardial interface, maintaining lower stimulation thresholds.
2. Fixation mechanism. It is necessary to secure the lead in the myocardium thus maintaining capture. Passive fixation is achieved with small flexible protrusions from the lead tip, *tines*, which hook into the trabeculated muscle. For less stable positions active fixation mechanisms are available, often a small retractable screw (Fig. 25.5D).
3. Lead conductor. This is the wire, often made of a complex alloy, which transmits the electrical current. Bipolar leads require two, one each for the cathode and the anode. Coiling the wires produces great lead flexibility, the anodal conductor being wound inside the cathode, separated by an insulator (*coaxial wire*). A recent advance involves coating each individual conductor with insulation (*coated wire technology*), which allows the anodal and cathodal conductors to

run together producing leads with even smaller diameters.
4. Lead insulation. Silicon was used initially; polyurethane is used today; it allows flexibility, resistance to damage, biocompatibility and good handling characteristics.
5. Lead connector. The metal connectors which link the conductor to the pacemaker. There is now an industry standard, *IS-1*.
6. Central lumen. A central lumen runs the length of the lead allowing the passage of a stylet that can be shaped to allow accurate positioning. A number of leads are also shaped to allow positioning in certain positions, e.g. the coronary sinus or the right atrial appendage.

Software

Pacemakers are becoming increasingly complex with multiple programmable parameters and functions, including:

1. Mode (see The NASPE/BPEG code, above)
2. Base rate – the minimum paced rate
3. Hysteresis rate – it is possible to allow the intrinsic heart rate to drop to a lower value, say 40, before pacing is initiated at a higher rate, say 60
4. Maximum paced rate – the maximum rate above which pacing will not occur, either driven by the patient's atrial rate or by the rate sensor
5. AV delay – the paced equivalent of the PR interval
6. Rate response functions. An increase in heart rate is a prime mechanism through which cardiac output is increased during both physical and mental activity. To allow a more physiological heart rate response in paced patients with abnormal sinus node function, many pacemakers sense this requirement; motion sensors, changes in transthoracic impedance as a marker of respiratory rate and QT interval are used. The pacing response can then be tailored to each patient
7. Sensitivity. The signal amplitude above which electrical activity will be 'seen'
8. Mode switch. Pacemakers should not track atrial tachycardias. Modern pacemakers can detect these arrhythmias and change mode accordingly.

The evolving complexity of the programmable parameters (the most important of which are summarized above) has made it increasingly difficult to diagnose pacemaker malfunction from the surface ECG without knowledge of the pacemaker set-up. In general terms, 'pacemaker malfunction' due to software or hardware failure is a rare event.

Future directions

Pacemaker therapy for traditional bradycardic indications is currently underused in the UK; implant rates are likely to increase. Furthermore, pacemaker indications are expanding beyond bradyarrhythmias, for example:

1. pacing to improve symptoms and prognosis in heart failure. Many patients with heart failure have conducting system abnormalities, e.g. left bundle branch block which produces dyssynchonous contraction of the right and left ventricles. Biventricular pacing (Fig. 25.5), in which both the right and left ventricle, via the coronary sinus, are paced, can restore co-ordinated ventricular activation and has been shown to improve symptoms and reduce mortality
2. pacing to prevent atrial tachyarrhythmia
3. pacing to prevent ventricular tachyarrhythmia in long QT syndrome
4. pacing to alleviate symptoms in vasovagal syncope
5. pacing to reduce the outflow gradient in hypertrophic cardiomyopathy.

In addition, we now have the ability to monitor pacemaker function remotely via wireless technology which will become increasingly important in years to come. Device-treated patients in the future are likely, therefore, to be part of an expanding, increasingly heterogeneous group requiring highly individualized care.

VAGAL NERVE STIMULATORS

Another application of pacemaker technology that is becoming increasingly prevalent is the implantable vagal nerve stimulator (VNS). This is currently approved as adjunctive therapy for medically refractive epilepsy and major depression. Possible indications in the future may eventually include obesity, chronic pain syndromes and various neuropsychiatric disorders, such as obsessive compulsive and panic disorders.

The system consists of a constant current pulse generator placed subcutaneously, as with cardiac pacemakers, in the pectoral region and connected onto the *left* vagus with a tunnelled lead culminating in spiral platinum electrodes embedded in silicone rubber (Fig. 25.6). The left nerve is used as the right vagus nerve carries a higher proportion of cardiac efferents. Again, as with cardiac pacemakers, subsequent programming is via an external computer with radiofrequency signals. Stimulation patterns are episodic (e.g. up to 90 s on and 5–10 min off) and patients carry a magnet whose transient application results in an additional pre-programmed burst of stimulation which may abate or prevent a seizure. The magnet can be held or fixed over the signal generator to pause stimulation, in order to limit some of the problems below.

Patients usually also carry a copy of the manufacturer's instructions for reference.

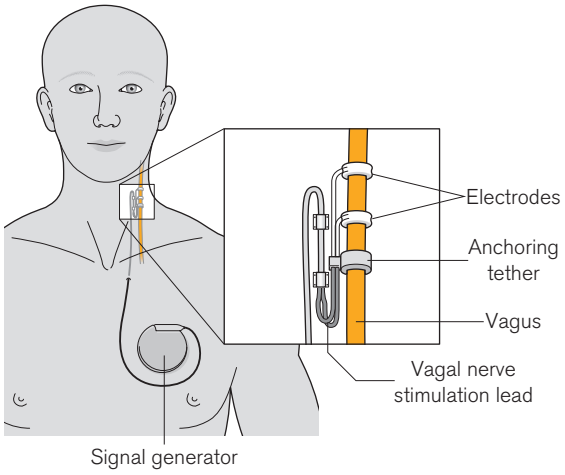


Figure 25.6 VNS system positioning.

In addition to the usual issues regarding electromagnetic interference (see later), there are a number of specific problems that are noteworthy for anaesthetists. They are mostly related to 'on time' of the device, when stimulation is taking place:

- Effects on heart rate and automaticity. Bradycardia appears commonplace; complete heart block and asystole are reported under anaesthesia and particularly at device insertion/activation.
- Vocal cord and laryngeal apparatus dysfunction, resulting in cough, voice alteration, swallowing difficulties and aspiration.[3]
- In addition to the above, effects on respiratory control (central and/or peripheral) are noted which can manifest as induced or worsened obstructive sleep apnoea.

DEFIBRILLATORS

Introduction

To understand defibrillator design and function and why defibrillation may fail requires some understanding of the pathophysiology of fibrillation and defibrillation.

Ventricular fibrillation (VF) is a complex arrhythmia consisting of random, disorganized, three-dimensional waves of depolarization, thus producing the loss of cardiac output and allowing arrhythmia persistence.

The mechanisms through which a shock of sufficient strength allows the return of spontaneous, co-ordinated electrical activity remain incompletely understood. A shock can have three effects depending on its timing in the action potential. Early on, during the absolute refractory period, no effect occurs. Later on, action potential prolongation is seen. Later still, a new action potential is induced. It is hypothesized that a shock of sufficient strength and appropriate timing will extend the refractory period in enough of the myocardium to allow the waves of depolarization to die out. Subsequently, cardiac automaticity allows the return of normal electrical activity, depending upon the underlying cardiac condition.

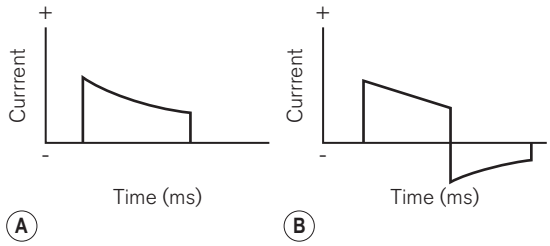


Figure 25.7 Diagrams showing the change of current with time for a monophasic (**A**) and a biphasic shock (**B**).

The amount of current required to produce defibrillation is known as the defibrillation threshold (DFT). For historical reasons, this is given in terms of energy (joules). Its determinants explain the success or failure of a shock. As with the pacing threshold, there is a minimum current below which defibrillation will not occur. Above a certain level, detrimental effects may occur, reducing the likelihood of success. Between these values success is most likely. However, as the waves of depolarization in VF are entirely random, an element of chance exists that the shock is delivered at the optimal time to defibrillate a critical mass of myocardium. This concept is known as the *probabilistic nature* of DFTs.

Factors affecting DFTs include:

1. Charge characteristics. A monophasic shock is one where the polarity of the shock remains constant throughout its delivery (Fig. 25.7A). In biphasic shocks (Fig. 25.7B) the polarity is reversed during delivery. In general terms, DFTs are lower and there is less post-shock myocardial depression with biphasic shocks. The shape and time-course of the shock can also be varied to produce optimal effects.
2. Electrode position. In implantable systems shocks can occur in a number of configurations that may affect the DFT (see Hardware, below). Altering the position of the endocardial coils can also have some effect. External paddle positions may also have an effect.
3. Shock polarity. Which electrode is used as the anode and which the cathode can affect DFTs, particularly with monophasic shocks.
4. Underlying cardiac condition. The more severe the cardiac condition, as assessed using heart size, ejection fraction, QRS width or heart failure symptoms, the higher the DFT. Furthermore, the

time spent in VF prior to defibrillation adversely affects DFTs.
5. Metabolic disturbance. DFTs are higher in hypoxic and acidotic patients.
6. Medication. DFTs can be affected by numerous medications. Of note, intravenous amiodarone reduces DFTs, whereas its chronic administration increases them. Fentanyl reduces DFTs. Common inhalational anaesthetic agents do not appear to have significant effects.
7. Shock impedance. Shock impedance affects current delivery to the myocardium. The transthoracic impedance for external defibrillation is dependent on lung volume, skin contact and tissue thickness.

External defibrillation

In the cardiac arrest situation and with external cardioversion for non-arrest arrhythmias, the energy is applied from the outside, positioned as for transcutaneous pacing. To improve current flow, gel or gel pads are used between the paddles and the skin. This allows better contact reducing impedance and the incidence of skin burns. Downward pressure on the paddles serves a similar purpose. Some defibrillator pads can also be used for monitoring and pacing. Smaller paddles for use during cardiac surgery are also available.

Current resuscitation recommendations recommend the use of 360 J monophasic or 150–360 J biphasic shocks for the defibrillation of VF and pulseless ventricular tachycardia. Biphasic shocks may allow the use of lower energies with less post-shock cardiac dysfunction. For the cardioversion of more co-ordinated, non-arrest arrhythmias, e.g. atrial flutter, it may be possible to use lower energies, whether mono- or biphasic.

The implantable cardioverter defibrillator

Implantable cardioverter defibrillators (ICDs) have been commercially available since 1985, enabling those at risk of recurrent life-threatening ventricular tachyarrhythmia to experience an improved quality of life. Over recent years technological improvements have allowed improved defibrillation success, easier implantation, greater longevity and increased programmability. Initially the devices were large and required abdominal placement and epicardial patches. In 2011, devices as small as 33 cm^3 are available and can be implanted in the same manner as pacemakers.

Hardware

The ICD system (Fig. 25.5) consists of a box containing:

1. Battery. The defibrillator battery has a number of requirements in common with pacemaker batteries:
   - **a.** small size
   - **b.** low self-discharge rate
   - **c.** predictable discharge rate
   - **d.** good longevity.

   However, it is also required to produce a large current over a short period to produce the shock, something the lithium-iodine battery is incapable of. The lithium silver vanadium oxide battery is used instead. Of note, because of its differing electrochemistry, ICD batteries are much less efficient for pacing, a feature reducing their longevity.
2. Capacitor. The ICD battery is unable, by itself, to produce sufficient charge quickly enough to allow defibrillation. The charge is, therefore, built up and stored in a capacitor which is discharged when required by completing the circuit between the capacitor plates via the myocardium. Modern ICD capacitors consist of thin films of aluminium, separated by a non-conducting layer of aluminium oxide
3. Voltage/output circuits. These allow different voltages to be produced for pacing and for capacitor charging. In addition the capacitor output can be controlled to produce the biphasic shock
4. Telemetry circuits
5. Pacing circuits
6. Sensing circuits
7. Memory
8. Reed switch.

Defibrillator leads are constructed in a very similar way to pacing leads with some added complexity:

1. A pace/sense electrode. This is essentially identical to a normal lead for a pacemaker. The cathode is found at the tip, with the anode either separate from or integrated with the distal shocking coil.
2. One/two shocking coils. The distal lead coil is positioned in the right ventricle and the proximal coil in the superior vena cava. They are constructed of similar material to pacing electrodes, but with a large surface area to improve current delivery. It is through these coils that the charge is distributed. A number of shock pathways are possible which may produce varying DFTs. The most common is between the coils and the ICD box; it is also possible to shock between the box and one coil or between one coil and the other.
3. A fixation mechanism.
4. Lead conductor. The conductors are of similar construction to those of pacing leads. With a dual coil, bipolar defibrillator lead four are necessary – one per coil and one each for the pacing anode and cathode
5. Lead insulation.
6. Lead connector. An IS-1 connector is used for the pace / sense functions. An industry standard connector for the defibrillator coils, *DF-1*, has been

defined. Standard ICD leads currently have three connections, one IS-1 and two DF-1 connectors, making them bulky. A new slimline connector incorporating all necessary connections on one lead is now available, (IS-4) and will be increasingly used.
7. Central lumen.

Software

The first generation of ICDs was only able to give therapy above a non-programmable ventricular rate, defined prior to device manufacture. As time has progressed, the complexity and programmability of the current third-generation devices has increased to give:

1. up to three zones for the detection and treatment of different ventricular tachyarrhythmias
2. multiple electrogram characteristics can be used to improve the sensitivity and specificity of differentiating ventricular arrhythmia from other rhythms (rate, onset characteristics, rate stability, ECG morphology, relationship of atrial to ventricular activity, etc.)
3. varying treatment options, which are programmable, including anti-tachycardia pacing, low-energy cardioversion and defibrillation
4. aspects of shock characteristics that are programmable including varying energies, shock pathways, polarity, timing, etc.
5. single, dual or biventricular pacing functions
6. in some devices, the ability to detect and treat atrial tachyarrhythmias
7. the opportunity for electrophysiological studies to be performed non-invasively.

Future directions

As time goes on and as our understanding of VF and defibrillation improves, it is likely that both external and internal defibrillation will change. Important changes in the short term will include the remote monitoring of ICD function as for pacemakers and a dramatic expansion in ICD indications and, therefore, in implant numbers and newer leadless ICDs (implanted subcutaneously, which may allow a greater proportion of the population to receive ICDs).

ELECTROMAGNETIC INTERFERENCE

Pacemakers are required to detect low amplitude electrical signals and communicate with pacemaker programmers with radiofrequency transmissions. They are, therefore, susceptible to interference from external sources of electromagnetic radiation. As pacemaker and ICD design has progressed and with the increased use of bipolar leads, electromagnetic interference is now less of an issue.

Frequencies of between 0 and 10^{11} Hz, representing radiofrequency and microwave energies, can affect pacing systems. Pacemakers respond to electromagnetic interference in different ways. If the device recognizes the signal as noise it can react by turning off its sensing circuits during the noise (VOO/DOO), resetting permanently to a default mode (often VVI) or by triggering a paced beat to prevent noise inhibiting its output. Clearly if the signals are not recognized as noise, pacemaker inhibition, reprogramming or ICD shock therapy may be triggered.

There are many sources of electromagnetic interference which can affect pacemakers and ICDs. However, those that are likely to influence anaesthetic practice include the following:

1. Diathermy – this can affect pacemakers in a number of ways, as discussed above. In addition, the diathermy current can activate the rate-response circuitry, damage the pacing circuits or pass via the lead damaging the lead-myocardial interface thus affecting pacing thresholds. Bipolar pacemakers are much less susceptible to this. Suggested management includes:
   a. preoperative consultation with the pacing technicians to enable programming alterations (asynchronous mode, rate response and magnet functions off, ICD tachyarrhythmia therapy off)
   b. the use of bipolar diathermy or unipolar diathermy as far as possible from the device, in short bursts only. Urgent pacemaker interrogation may be required. A magnet can be used in an emergency, but may, in some circumstances, start automatic threshold testing or open the device to reprogramming
   c. post-surgery a further pacemaker test is recommended.
2. Cardioversion/defibrillation – like diathermy, the considerable current used can overwhelm the protective circuitry and reprogramme or damage the device or the lead–myocardial interface. Management includes device interrogation pre- and post-procedure. Paddles should be placed at least 10 cm from the box and at 90° to the axis of box to lead tip (preferably AP).
3. MRI – the scanner can interact in a number of ways with pacing systems, not least because of the ferromagnetic components of old devices. Strong electrical signals are also produced that can either inhibit or induce rapid pacing. In general terms, an MRI is contraindicated in pacemaker/ICD patients, although case reports exist of patients undergoing this investigation without ill-effect. There is now an MRI-safe pacemaker available from Medtronic with the other pacing companies soon to follow suit.

The model is safe to use in an MRI scanner six weeks after implant. MRI-safe ICDs are also likely to come at some point.

4. Lithotripsy – the shock can trigger an arrhythmia, cause mode switch or pacemaker inhibition or damage the box and, in particular, the piezoelectric crystals used for activity sensing. The manufacturer should be consulted particularly for abdominally placed systems, devices should be programmed to an asynchronous mode and ICDs should be disabled.
5. Radiotherapy – if directed at the pacing box this may destroy the circuitry. Pacing systems require repositioning.
6. Electroconvulsive therapy – adverse effects are unlikely, but pacemaker checks and asynchronous pacing are suggested.

In general terms, consultation in advance with the pacing technicians for up-to-date advice is strongly recommended. Magnet placement may not be sufficient and may even be detrimental.

REFERENCES

1. McEneaney DJ, Cochrane DJ, Anderson JA, Adgey AA. A gastroesophageal electrode for atrial and ventricular pacing. *Pacing Clin Electrophysiol* 1997;**20**:1815–25.
2. Andersen HR, Pless P. Trans-esophageal pacing. *Pacing Clin Electrophysiol* 1983;**6**:674–9.
3. Lundgren J, Ekberg O, Olsson R. Aspiration: a potential complication to vagus nerve stimulation. *Epilepsia* 2005;**39**:998–1000.

FURTHER READING

Cummins RO, Hazinski MF, Kerber RE, Kudenchuk P, Becker L, Nichol G. Low-energy biphasic waveform defibrillation: evidence-based review applied to emergency cardiovascular care guidelines. *Circulation* 1998;**97**:1654–67.

Ellenbogen KA, Wilkoff BL, Kay GN, Lau CP. *Clinical cardiac pacing, defibrillation and resynchronization therapy*. 3rd ed. Philadelphia: WB Saunders; 2006.

Epstein AE, DiMarco JP, Ellenbogen KA, Estes NA 3rd, Freedman RA, Gettes LS, et al. ACC/AHA/HRS 2008 guidelines for device-based therapy of cardiac rhythm abnormalities. *J Am Coll Cardiol* 2008;**51**:e1–62.

European Resuscitation Council Guidelines 2005 for Resuscitation. *Resuscitation* 2005;**67**:S1.

Gammage MD. Temporary cardiac pacing. *HEART* 2000; **83**:715–20.

Gold MR. Permanent pacing: new indications. *HEART* 2001;**86**: 355–60.

Hatton KW, McLarney JT, Pittman T, Fahy BG. Vagal nerve stimulation: overview and implications for anesthesiologists. *Anesth Analg* 2006;**103**:5,1241–9.

Murphy JJ. Problems with temporary cardiac pacing. *Anesth Analg* 2001;**323**:527.

http://www.physiocontrol.com/learning/clinical-topics/defibrillation.aspx. Accessed on 2nd Aug 2011.

http://www.physio-control.com/learning/clinical-topics/index.aspx?id=2147483878. Accessed on 2nd Aug 2011.

Chapter 26

Lasers

Patrick T Magee

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PRINCIPLES

The increasing surgical use of lasers, with their inherent potential hazards to patients and operating room staff, mandates an understanding of their physical principles by anaesthetists.[1]

The word *laser* is an acronym for 'light amplification by stimulated emission of radiation'. The laser produces an intense beam of pure monochromatic light (one wavelength: one colour), in which all of the waves are in phase (coherent). The output beam is likely to be of a very small cross-sectional area and is virtually a non-divergent parallel (collimated) beam. These properties mean that energy may be delivered to very small areas of tissue with great accuracy, and the intense parallel beam of light constitutes a very large amount of power per unit area of tissue. The wavelength of a laser is determined by the lasing medium used. Although described as monochromatic, most laser media produce light within a narrow waveband consisting of a number of discrete frequencies. In order to come up with the concept of a laser, which was first described in 1958 and first demonstrated in 1960, scientists had to understand the notion of quantum physics and of Niels Bohr's model of the atom, with its orbital discrete energy levels (see below).

Although there are many more complex laser systems,[2] the basic components of a laser are shown in Fig. 26.1. The lasing medium may be a solid, liquid or gas. The atoms of a lasing medium are excited to high energy levels by a 'pumping' source, which may be a high-voltage discharge in the case of a gas, an intense flash of light from a flashtube or the energy from a radio frequency power source. Fig. 26.2 shows the excitation and emission process possible in a gaseous lasing medium. A photon of energy from the pumping source may be absorbed by a stable atom in its so called 'ground state', which then becomes an atom in an excited state, with an electron or electrons in an orbital shell at a higher energy level. Spontaneous emission of a photon of energy occurs as the electrons fall back to shells of a lower energy state and the excited atom reverts to the ground state.

If a further photon of pumping energy, at the correct wavelength, is applied to an atom in its excited state, it will fall to its ground state and two photons of energy will be emitted instead of one. This is known as stimulated emission, originally described by Einstein in 1917 as the basis for laser technology[3] and the inversion of the energy states is referred to as *population inversion*. The emitted photons thus produced are in phase with, have the same polarization, and travel in the same direction, as the stimulating radiation. This mechanism is amplified by many of the escaping photons being reflected back into the lasing medium by the mirrors. Thus a chain reaction occurs, and this can be thought of as a positive feedback system. The process produces an intense source of light energy, some of which is allowed to escape through the partially reflecting mirror at the output end of the lasing medium. The output beam of the laser is usually directed to the tissues through a fibre-optic light guide. However, the wavelength of the carbon dioxide laser is so long, at 10.6 µm, that there is no fibre currently available to transmit energy, so that it has to be directed by a series of mirrors instead.

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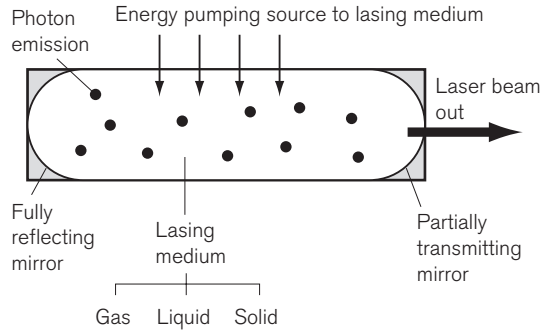


Figure 26.1 Basic components of a laser.

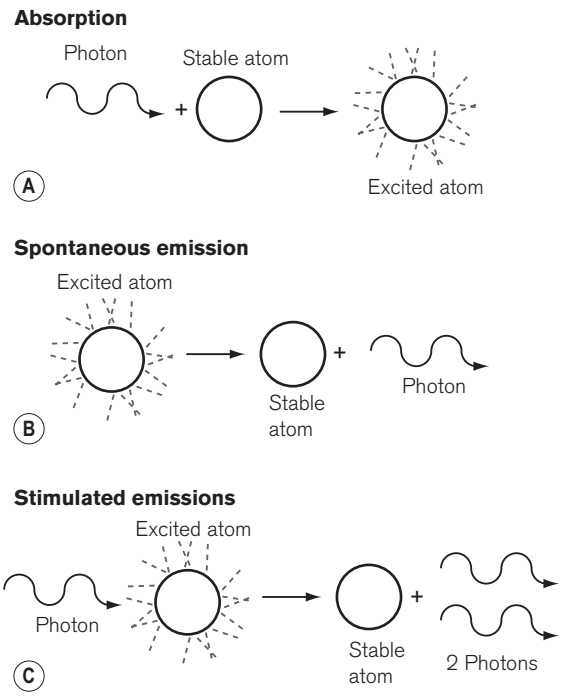


Figure 26.2 Absorption, excitation and emission processes.

CLINICAL APPLICATIONS

The clinical use of lasers depends on a compromise between:

- laser–tissue interaction[4]
- absorption and penetration depth
- availability of a laser of the correct wavelength and power
- availability of a suitable method of transmission.

The extent of the effect of a laser on human tissues depends primarily on the intensity and the frequency of the light being used.[5] At low light intensity, stimulation occurs within the cell and this is exploited in physiotherapy applications. At slightly higher intensity, attenuation of cellular activity occurs. At still higher intensity levels, about 40 J cm^{-2}, sensitizing agents in tissue become activated (this is the basis of protecting the skin from the sun's ultraviolet rays using suntan lotions). By the time the light intensity has risen to 400 J cm^{-2}, the tissue temperature has risen to 60°C and protein denaturation and photocoagulation predominate. Further large increases in light intensity result in a tissue temperature rise to 100°C, vaporization of tissue fluids and destruction of cell structures. In order to control the destructive power of a laser, most systems can be pulsed; the light is emitted in short bursts, to allow heat dissipation between bursts and reduction of thermal damage to neighbouring tissues. *'Q switching'* of a laser refers to a device which allows aliquots of laser light to be stored and released in bursts of even higher energy and shorter duration. Such a technique is used in ophthalmology to cause photoablation and to minimize thermal damage to the eye. With Q switching, while collateral thermal damage may be reduced, the frequency of switching is such that tissue vibration and, therefore, mechanical damage may predominate.

The penetration of light energy into body tissues depends on the wavelength of the light. Far infrared and ultraviolet light has little penetration because it is rapidly absorbed near the surface of the tissue, by tissue water. Maximum penetration of light occurs at the red end of the spectrum.

The graph in Fig. 26.3 shows the spectrum of absorption of light at different wavelengths by haemoglobin, melanin and water. Monochromatic light energy is absorbed by tissue of *complementary colour* (opposite colour) and reflected by substances of the same colour.

Carbon dioxide laser energy at 10.6 µm is absorbed by water within 1 mm depth, causing rapid vaporization of intracellular water. The main use of the carbon dioxide laser is, therefore, as a bloodless cutter and vaporizer. The blue-green argon laser beam penetrates to about 2 mm depth and is maximally absorbed in the 500 nm waveband by substances of a complementary colour (red), such as haemoglobin. Thus, the argon laser is used to coagulate blood in small vessels with very little effect on other more transparent tissues, for example the retina. The Nd:YAG (neodymium: yttrium-aluminium garnet) laser has a solid lasing medium and produces energy in the near infrared region of the spectrum, which has maximum penetration, being absorbed at 3–5 mm depth, by haemoglobin, melanin and water. When invisible infrared lasers are used (e.g. CO_2 laser), it is common practice to make use of a low powered, visible-light laser, such as a helium:neon laser at the same time, in order to aid in aiming the therapeutic laser accurately. Table 26.1 shows the currently available lasers in medical use.

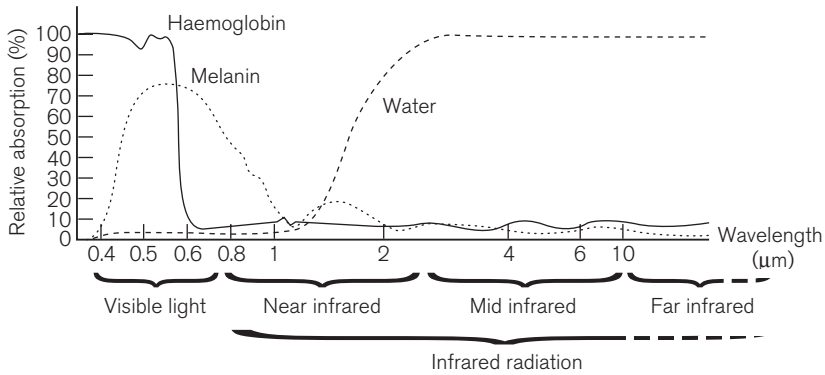


Figure 26.3 Absorption characteristics of tissue constituents.

Table 26.1 Some currently available medical lasers

Lasing Medium	Wavelength, nm	Colour	Transmission
Krypton	476, 521, 568, 647	Blue to red	Optical fibre
Argon	488–515	Blue-green	Optical fibre
Nd-YAG-KTP	532	Green	Optical fibre
Helium-neon	633	Red	Optical fibre
Nd-YAG	1064	Near IR	Optical fibre
CO_2	10600	Far IR	Mirrors

SAFETY ASPECTS

Apart from the danger to the patient from the beam of laser energy if it is misused, there is a risk to the operator and other persons in the operating environment.[6] This is because of the long range of laser light due to the virtual non-divergence of the beam; thus, in contrast to a collimated X-ray beam for example, increased distance from the source has very little safety benefit. Even reflected laser light may be very dangerous to the eyes. Visible laser light transmitted to the retina of the eye may burn it irreparably, leaving a blind spot in the field of vision. A similar lesion over the optic nerve may result in total blindness of that eye. The cornea, lens and aqueous and vitreous humours partially or totally absorb far-infrared laser radiation; these tissues, therefore, are more susceptible to damage than the retina.

Laser radiation on the skin may be felt as a burning sensation, which is, therefore, self-protective, provided that the victim is conscious and has not received analgesia.

In terms of the danger that lasers pose to humans, there is a complex relationship between power, frequency and time of exposure. There is an international classification for lasers, shown in Table 26.2, where Class I lasers are inherently safe and Class IV lasers are, broadly speaking, hazardous if misused. Most lasers in medical use are in Class IV. No one should use a laser who is not trained to do so, and everyone who is working in the vicinity of a laser should be trained in the safety aspects of its use. This includes the anaesthetist, who is often standing in the line of fire of the laser.

Table 26.2 International classification of continuously working lasers

Class I	Powers not to exceed maximum permissible exposure for the eye
Class II	Visible laser beams only; powers up to 1 mW; eye protected blink-reflex time of 0.25 s
Class IIIa	Relaxation of class II to 5 mW for radiation, provided beam is expanded so that the eye is still protected by the blink-reflex
Class IIIb	Powers up to 500 mW; direct viewing hazardous
Class IV	Powers over 500 mW; extremely hazardous

Chapter 27

Provision of anaesthesia in difficult situations and the developing world

Deborah Harris and John A Carter

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The provision of anaesthesia in modern well-equipped operating theatres is dependent on sophisticated electronic equipment that requires an uninterrupted supply of both electricity and compressed gasses. Such equipment is not readily transportable, although it may be moved within a hospital facility. There are many locations throughout the world where anaesthesia is administered to facilitate surgery, investigations or other forms of treatment outside this generally accepted 'safe' environment.

The following are examples of locations and situations away from hospital operating theatres where anaesthesia may be required, and where simpler or alternative means of providing anaesthesia may need to be employed:

- within hospitals away from operating theatres:
  - Accident and emergency departments
  - Radiology departments
  - Magnetic resonance imaging suites
  - Radiotherapy departments
  - Intensive care units
  - Coronary care units – e.g. for cardioversion
  - Psychiatric Units – for electroconvulsive therapy
- interhospital transfers
- in developing countries:
  - Hospitals, medical centres
  - Self-contained visiting surgical teams
- at the site of an accident or major disaster
- on the battlefield.

Domiciliary anaesthesia – as in kitchen table appendicectomy and obstetric flying squad interventions – has long been abandoned on safety grounds and, more recently, so has anaesthesia in dental surgeries in the UK.

All of these situations are remote from the relatively safe, comfortable and familiar operating theatre anaesthetic environment, and the following problems may be encountered to a greater or lesser degree:

- lack of continuous electricity supply
- lack of continuous supply of oxygen and nitrous oxide
- difficulty with storage of drugs and equipment
- difficulty in transport and supply of drugs and equipment
- lack of maintenance of equipment
- lack of skilled assistance
- lack of control of environment
- financial restrictions.

Where possible, on grounds of safety, patients should be transferred to medical facilities capable of providing the appropriate level of care. For example, electroconvulsive therapy for the psychiatric patient with severe aortic stenosis and depression would be better managed (from their cardiac status) in the operating suite of the main hospital rather than in a room off the psychiatric ward.

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Non-essential surgery should not be undertaken at the site of a major disaster or on the battlefield, and the use of local, regional or sedative techniques should be considered where appropriate.

The overriding principle in providing anaesthesia under any of these conditions should be to use a simple, safe technique familiar to the practitioner. To reduce complexity and avoid the potential administration of a hypoxic gas mixture as well as reducing the need for scavenging (and for many other well-documented reasons), there is a case for avoiding the use of nitrous oxide entirely. Training and practice in such techniques is invaluable for the time when they may be required. Even within a modern operating theatre environment, a 'difficult situation' may arise due to failure of a sophisticated electronic anaesthetic workstation, a major power cut with failure of back-up generators or a disruption to piped gas supply. The use of total intravenous anaesthesia (TIVA) together with a self-inflating bag and a free-standing oxygen cylinder, combined with practical clinical monitoring, will allow adequate and safe anaesthesia in such a situation. Under such circumstances a hands-free torch or headlight may be the most essential item of additional equipment.

DIFFICULT SITUATIONS WITHIN HOSPITALS

Sites away from the operating theatres often have anaesthetic equipment that is used only occasionally. Piped oxygen and suction facilities may be absent. The equipment in such areas must be maintained and checked adequately, with basic monitoring meeting the standard recommended by the Association of Anaesthetists.[1] Since January 2003, all anaesthetic machines in use in the UK must be incapable of delivering a hypoxic mixture. There must be immediate access to resuscitation equipment and drugs, and a means of summoning additional assistance (i.e. telephone or intercom). The anaesthetist and their assistant should have sufficient experience and be familiar with both the environment and the equipment.

Some specific problems with regard to patients, medical attendants and equipment within particular areas are listed below.

Radiology departments

- Ionizing radiation risk
- Long procedures – e.g. coiling of intracerebral aneurysms
- Low levels of lighting
- Restricted access to patient or patient's head
- A possible requirement to stop ventilation briefly to avoid image blurring.

Radiotherapy units

- Intense ionizing radiation requiring patient isolation from the medical attendants
- Closed circuit television or glass-liquid-glass window to view patient causing colour and image distortion
- Multiple frequent treatments over a few weeks
- Radiotherapy applicators may obstruct access to the patient's head.

Magnetic resonance imaging (MRI)[2]

- Intense magnetic field with the ability to cause equipment made of ferromagnetic material to be attracted at projectile velocity into the scanner. There is, however, a rapid decrease in field strength with distance
- Electrical inductance – potential for thermal injury from electrical conducting leads
- Electromagnetic interference leading to equipment malfunction (e.g. in syringe drivers) or artefact (e.g. altered ST-T region on ECG, from electrical currents induced by aortic blood flow in a magnetic field, rendering ischaemia detection difficult)
- Noise from vibration of switched gradient coils makes audible alarms inappropriate and necessitates ear protection
- Theoretical risk of hypoxia if quenching of the super-conducting magnets of cryogenic gasses (usually helium) occurs. Quenching may occur as a fault condition or be initiated for emergency shutdown of the magnet, and should be safe if gasses are appropriately vented to the outside.

These factors pose risks to patients and potential occupational hazards to staff. Patients and staff must be screened before access is granted to an MRI scanner, to exclude ferromagnetic implants, such as aneurysm clips or pacemakers.[2] Anaesthetic equipment taken into the vicinity of the MRI scanner must be MR-compatible.

Remote anaesthesia

Anaesthesia for MRI, radiotherapy and some radiological procedures may necessitate the anaesthetist and the bulk of the anaesthetic equipment being remote from the patient. This may be either to ensure all ferromagnetic equipment is outside the magnetic field, or to remove anaesthetic personnel from ionizing radiation:

- TIVA may be employed using long infusion lines on pumps which must be able to cope with the high resistance to flow caused by the increased length. This usually means setting to maximum the pressure limit for sensing an occlusion.
- Whilst sedation may be sufficient for some patients, the airway may need to be established with a supraglottic airway device or tracheal tube.

- Intermittent positive pressure ventilation through a long coaxial breathing system such as a 9.6–10 m Bain circuit and Nuffield Penlon series 200 ventilator, has been shown to provide safe anaesthesia.[3] With this system, there is an increase in the static compliance in proportion to the length of the tubing. This is caused by expansion of the breathing hose and compression of the volume of gas during positive pressure ventilation and will result in a lower tidal volume being delivered than is set on the ventilator although this may be mitigated by the tidal volume supplementation from the fresh gas flow. Capnography is essential. In children, if a Newton valve is used, the ventilator becomes a pressure generator, and the increased resistance and compliance of the long system results in the pressure delivered being significantly less than that selected (23% less with a 10 kg child). This compares to a 6–11% reduction when using a long rubber Ayre's T-piece.[4]
- The capnography signal is delayed due to the length of the sampling line but provides a guide for adjustment of the tidal volume.

INTERHOSPITAL TRANSFERS

It is sometimes necessary to transfer anaesthetized patients to another hospital, particularly if they require specialist services which are not available on site. Often these patients will be critically ill and the keys to their successful transfer are *communication*, *documentation* and *anticipation* of possible problems. All hospitals should have a checklist for interhospital transfers:

- Communication between transferring and receiving teams and the ambulance service is essential. It is imperative that everyone concerned knows the current condition of the patient, the indication for transfer and the exact destination. It is also the responsibility of the referring team to arrange a suitably qualified medical escort.
- Documentation must stay with the patient. This includes the notes from the whole patient episode plus any additional relevant past notes, blood results and X-rays – either as hard copies or on CD. If blood has been cross-matched, it is worth transferring with the patient if the transfer is less than 4 hours because even though the receiving hospital will want to reissue it, this will be quicker than starting a cross-match from scratch.
- The patient should be stabilized as far as possible prior to transfer. The accompanying anaesthetist is unlikely to be familiar with the ambulance or what sort of equipment is immediately available and, therefore, should plan for all potential problems. In particular they should calculate the amount of oxygen that is going to be required and whether that has been catered for, the battery life on any infusion pumps should be adequate to allow for delays in traffic, etc., and spare infusions drawn up if necessary. Similarly, basic checks of equipment, such as availability of endotracheal tubes, working laryngoscopes, self-inflating resuscitation bag and IV cannulae, should not be omitted.

DEVELOPING COUNTRIES

There are two extremes of conditions that may be met in providing anaesthesia in developing countries. The anaesthetist may be totally dependent on the equipment, drugs and personnel provided within the healthcare system of that country, or they may be part of a visiting team that is totally self-contained. Visiting teams may be very *operation specific* (e.g. Project Orbis, Operation Smile and other eye or cleft palate teams), or they may have a much wider remit. Operation specific teams usually have rigid pre-assessment protocols, ensuring that standardized procedures are carried out on fit patients, enabling the greatest good to be done for the largest number of people. Some visiting teams may bring all facilities needed to perform a certain number of specified operations and anaesthetics. The devices in use can then range from those seen in modern developed economies through to equipment similar to that used for battlefield anaesthesia (see below). Others opt to mainly use local equipment, adding only their own disposable equipment.

'District hospital'-based anaesthesia

Many small hospitals in developing countries rely on non-medically qualified assistants to deliver anaesthesia under the supervision of the doctor who will also be performing the surgery. Under these conditions, anaesthetized patients are more likely to be intubated to ensure a secure airway. Most anaesthetists in developing countries work in larger hospitals, but even here they may be responsible for the training and supervision of medical assistants giving anaesthesia. Many such hospitals, large and small, will have storerooms which have become graveyards of anaesthetic machines and other equipment donated by well-meaning organizations or countries, without consideration for the spare parts or expertise needed for their maintenance. There will often be continuous flow (Boyle's) machines, discarded as the necessary compressed medical gas supply is absent or erratic. In addition, such machines may not have anti-hypoxia devices and vaporizers may be outdated, unserviceable or grossly inaccurate.

For all these reasons, local anaesthetic techniques (nerve blocks, spinals and epidurals) should be used where appropriate.

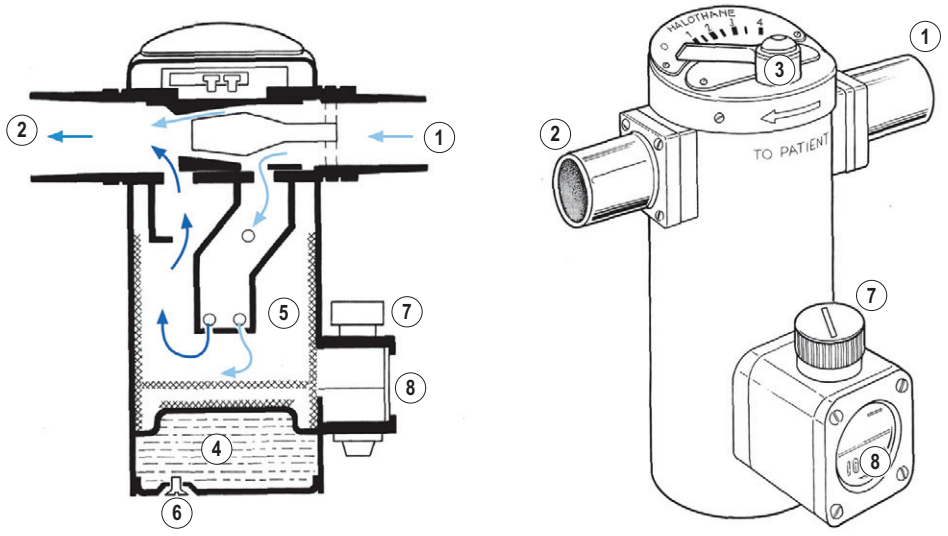


Figure 27.2 Oxford miniature vaporizer (OMV). (1) Inlet port, (2) outlet port, (3) concentration control, (4) heat sink (5) vaporizing chamber, (6) filling port for water, (7) filling port for anaesthetic, (8) anaesthetic-level indicator.
Reproduced with permission of WHO from Dobson MB. Anaesthesia at the District Hospital. *Geneva: World Health Organization; 1988.*

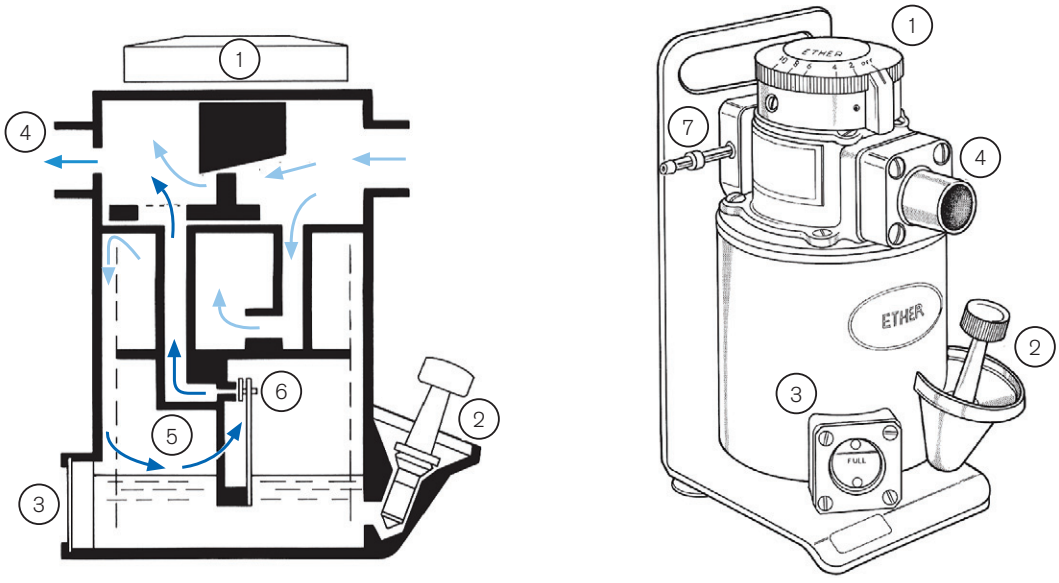


Figure 27.3 PAC (Portable Anaesthesia Complete) vaporizer (Ohmeda). (1) Concentration control, (2) filling port for ether, (3) ether-level gauge, (4) outlet and one-way valve, (5) vaporizing chamber, (6) thermocompensator valve, (7) port for oxygen enrichment.
Reproduced with permission of WHO from Dobson MB. Anaesthesia at the District Hospital. *Geneva: World Health Organization; 1988.*

available care. The issue of standards is in fact a significant and wide-ranging one. It is argued that the standards organizations, in being dominated by manufacturers, produce standards geared to driving sales of the latest technology to wealthy nations.[6] Such equipment does not address the needs of developing nations and the standards inhibit 'low technology' developments.

There are, however, courses available to demonstrate the use of such draw-over apparatus, using highly sophisticated anaesthetic simulators.

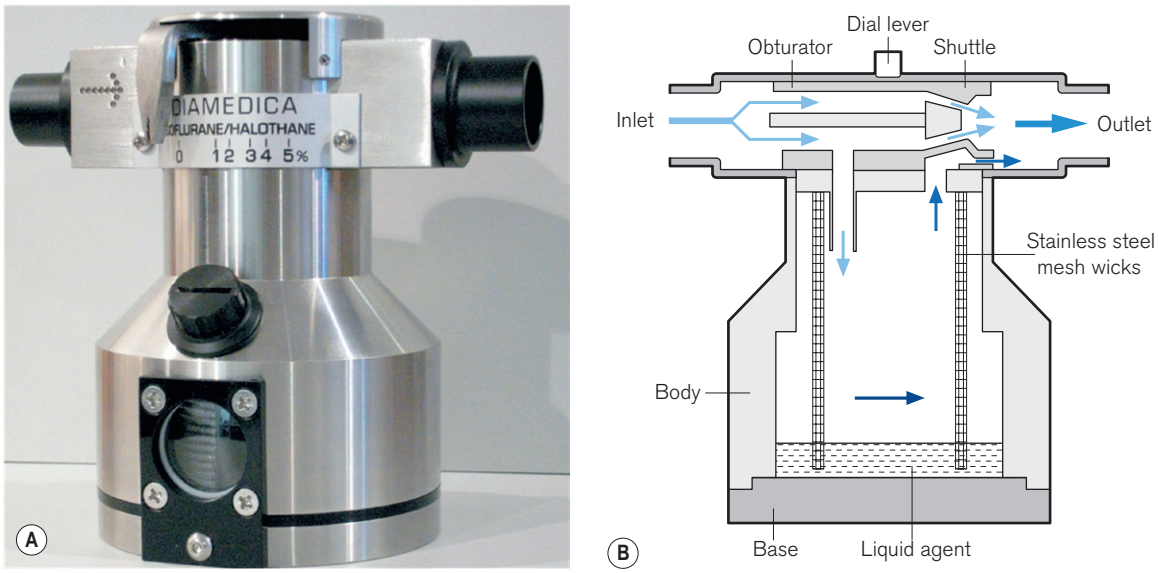


Figure 27.4 Diamedica vaporizer. This is a 150 ml agent capacity draw-over vaporizer for isoflurane or halothane. The dial lever causes a shuttle to move against the stationary obturator, altering the size of the bypass route annulus. A calibration nut (not drawn) moves the obturator and is for factory or service technician use.
Photograph courtesy of Diamedica (UK) Ltd. Schematic redrawn from an original kindly supplied by Diamedica (UK) Ltd.

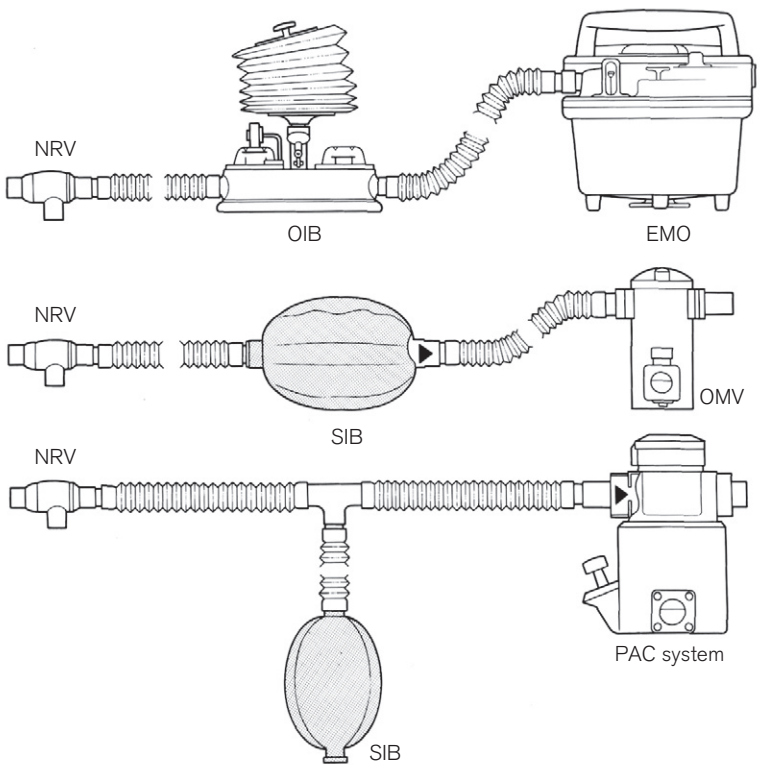


Figure 27.5 Several arrangements of draw-over apparatus: OIB, Oxford inflating bellows (Penlon); EMO, Epstein-Macintosh-Oxford ether vaporizer (Penlon); OMV, Oxford Miniature Vaporizer (Penlon); NRV, non-rebreathing valve (e.g. Ruben); SIB, self-inflating bag (Laerdal, Ambu, etc.); PAC, Portable Anaesthesia Complete (Ohmeda).
Reproduced with permission of WHO from Dobson MB. Anaesthesia at the District Hospital. *Geneva: World Health Organization; 1988.*

be unreliable, it is advisable to have manual or foot-operated suction apparatus.

Universal Anaesthesia Machine (UAM)

At the time of writing, the very newly produced UAM (Fig. 27.8) is starting to undergo field evaluations, and represents the only CE marked anaesthesia machine designed with the needs of developing countries in mind. Developed by Paul Fenton and manufactured by OES Medical (Abingdon, UK), production has been made possible as a result of funding by the Nick Simons Foundation: a charitable organization dedicated to providing medical care in rural Nepal.

The UAM keeps the failsafe features of draw-over anaesthesia, but by removing the need for a non-rebreathing valve at the patient end of the breathing attachment allows use of modern lightweight coaxial or Y configured dual-limb breathing tubes, which also facilitate waste anaesthetic gas scavenging. The key feature in this is the balloon valve, incorporated into the machine, which occludes the expiratory limb of the breathing system through an actuator pipe that is pressurized from the breathing system to allow positive pressure ventilation (Fig. 27.9). A one way flap valve distal to the balloon occluder ensures unidirectional flow through the breathing system during spontaneous respiration.



Figure 27.8 Universal Anaesthesia Machine.
Photo courtesy of OES Medical, Abingdon, UK.

The machine houses a built-in oxygen concentrator capable of producing up to 10 L min^{-1}, but also has cylinder yokes for oxygen and nitrous oxide as well as connections for pipeline oxygen. An additional low pressure O_2 inlet is provided distal to the flowmeters. A fuel cell oxygen analyzer downstream of the calibrated vaporizer linked to an electronically operated valve shuts off the nitrous oxide supply where a minimum of 35% O_2 is not detected. Positive and negative pressure relief valves upstream of the vaporizer and reservoir bag allow entrainment of room air and hence anaesthesia delivery in the absence of power or gas supplies. A pressure transducer at this level can act as an 'apnoea alarm' by warning of distension of the reservoir bag, which indicates diminished patient minute volume in comparison to the fresh gas flow rate. The inflating bellows can be used to assist or take over ventilation as in other draw-over systems.

Future iterations of the UAM are planned to include an in-line electrically powered bellows ventilator within the oxygen concentrator housing and an integral circle system and absorber. The aforementioned, with high-quality production values and materials, multiple source oxygen facilities allowing continuous flow gas delivery, low maintenance requirements, the use of modern breathing systems in a draw-over configuration and CE marking; all in a restricted cost device, aim to permit use in any hospital setting.

MAJOR ACCIDENTS AND DISASTERS

These may occur in any part of the world at any time, and are by definition unexpected. All medical services should have a plan to deal with major disasters. A typical approach is to have a mobile medical team that can be rapidly deployed to the disaster site and a receiving hospital capable of dealing with the retrieved casualties. In the event of the number of casualties overwhelming the initial response, there should be a means of either escalating the number of teams or hospitals deployed. In developing countries, there may be a need to seek international assistance. Many countries have teams available for worldwide deployment at short notice. Particular problems encountered include:

- unfamiliar territory
- unfriendly environment:
  - extremes of hot and cold and altitude, even in normally temperate climates
  - dark, wet, cramped conditions
- unfamiliar injuries:
  - blast and crush injuries
  - delayed extrication
- risk to rescuers:
  - nuclear, biological or chemical incidents
  - terrorism
  - fire, explosion risk

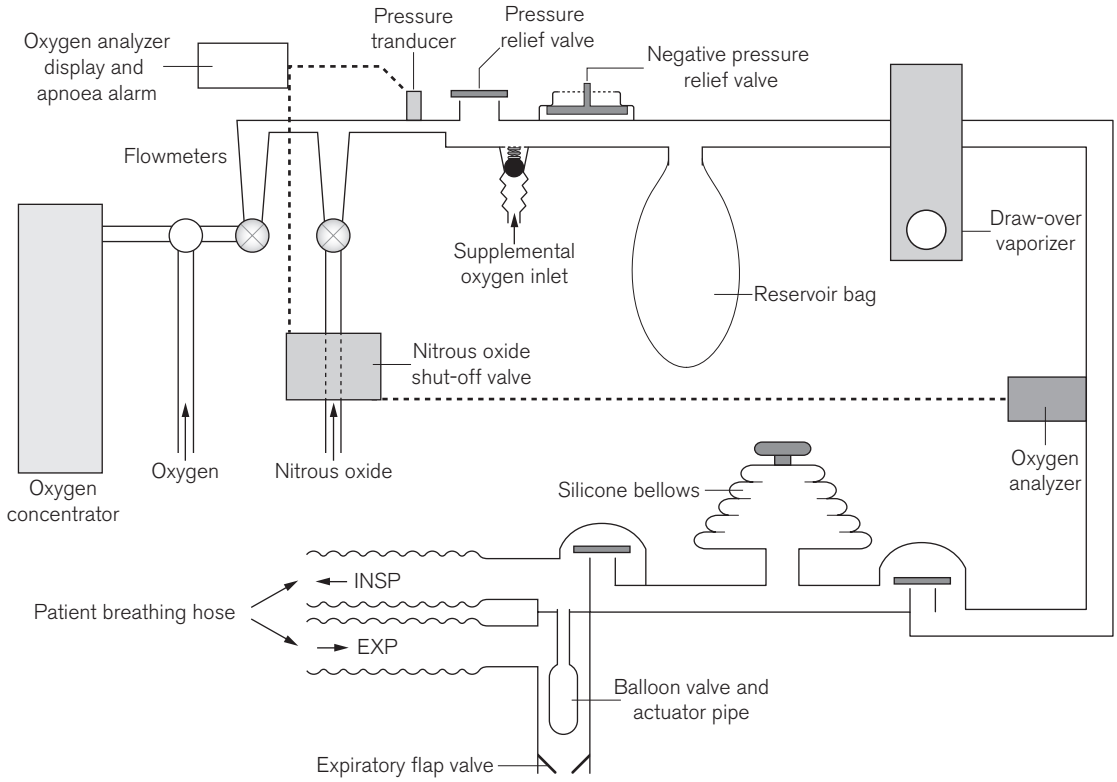


Figure 27.9 Gas flow schematic of the UAM from a drawing by Dr Paul Fenton at www.uamachine.org. See text.

- continuing disaster (e.g. earthquake)
  - unstable buildings.

The predominant anaesthetic contribution to a major disaster is resuscitation and stabilization prior to transfer to the receiving medical facility. Exceptionally, to aid extrication of casualties, amputation of trapped limbs may be required. This is best achieved using ketamine, either intravenously or intramuscularly. Equipment for intubation, self-inflating bag, fluid and cannulae for intravenous fluid resuscitation should be available. Oxygen and Entonox should be used cautiously in such conditions as they support and accelerate combustion of flammable materials, and whilst the latter has excellent analgesic properties, it may not be suitable in very cold conditions (see note Table 1.4) or in the presence of a head injury or pneumothorax.

THE BATTLEFIELD

Mobile field hospitals are deployed as close to the battlefront as safety will allow, and receive casualties who will normally have had only life-saving first-aid treatment at a regimental aid post or equivalent. In addition to military casualties from both sides of the conflict, there are frequently civilian casualties, which may include children. This poses a problem if paediatric equipment is not available. Large numbers of casualties may arrive simultaneously, and require triage on arrival. In some, immediate surgery is required as part of the resuscitation process. Some of the features of military anaesthesia are as follows:

- equipment must be air portable
- oxygen cylinders or concentrators are usually readily available
- cost constraints for drugs and equipment are minimal
- resupply is not usually a problem – unless supply lines are cut.

Electricity is required for lighting, monitoring, suction, heating and refrigeration (for blood storage and some drugs), and will usually be supplied from generators that must be of sufficient power to cope with maximum demand. Vital equipment should have an independent battery back-up. Many modern pieces of equipment have a back-up supply of only 10 min. Sensitive equipment should have surge protection to limit voltage spikes from erratic power supplies.

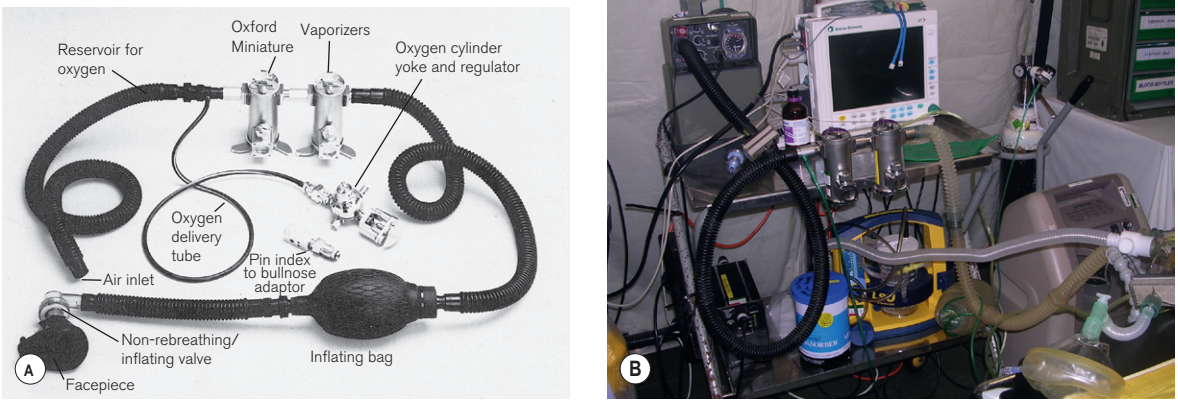


Figure 27.10 **A.** The Triservice apparatus. **B.**Triservice apparatus in use in push-over mode, Basra, Iraq (2007).
Photograph courtesy of Colonel S Jagdish LIRAMC, 33 Field Hospital.

Triservice apparatus

The equipment in use at the moment by the British military medical services is the Triservice Anaesthetic Apparatus (Fig. 27.10).[10,11] This consists of two modified OMVs connected via a self-inflating bag to a non-rebreathing valve and facemask or airway device. Supplemental oxygen may be delivered upstream of the vaporizers by a T-piece with a length of corrugated tubing acting as a reservoir. This complete apparatus, including an oxygen regulator and cylinder yoke, comes securely packed in foam within an air portable container, all weighing less than 25 kg, and can be safely dropped by parachute.

The OMV used in this apparatus has been modified by incorporating three folding feet to enable them to stand on a flat surface. Additionally, the capacity of the chamber has been increased to 50 ml. The wicks within the vaporizing chamber are of metal gauze, so that a different agent may be used by simply draining the vaporizer and rinsing the chamber (with a little of the new agent, which is then discarded), before properly charging the vaporizer for use.

Detachable calibration scales are supplied for different agents. When the control is turned to '0' (off), the contents will not spill if the vaporizer is accidentally inverted, although it is recommended that the vaporizer should be drained for transport. If it is tipped or inverted during use, the vaporizer must be kept upright for a few minutes before use to allow agent that may have entered the bypass or the control mechanism to drain back into the chamber, otherwise very high concentrations of vapour may be initially delivered. The OMV is not temperature-compensated, and although its body acts as a small heatsink, the vapour output concentration will decrease with time. By having two OMVs in series, it is possible to switch between them as the output from one starts to fall off, to switch between different agents, or to use both in series to deliver a higher concentration for induction of anaesthesia. Previously, trichloroethylene was administered alongside halothane to make up for the absence of the analgesic effect of nitrous oxide.



Figure 27.11 Pneupac compPAC Ventilator, Smiths Medical, UK.
Photo courtesy of Smiths Medical International.

Pneupac compPac ventilator

The Triservice apparatus may be used in spontaneously breathing patients, or IPPV may be instituted either using the self-inflating bag or by replacing the bag with a suitable ventilator. Originally the CapeTC50, a relatively portable ventilator consisting of a bellows expanded and contracted by an electric motor, was used. This has been replaced in British military use by the Pneupac compPAC ventilator (Fig. 27.11). This is a rugged, portable, gas-

powered ventilator which can be driven from an external gas source (3–6 bar) or from its internal compressor. An integral rechargeable battery or external 24/28V DC supply drives the compressor and there is the facility for admixture of oxygen from a low pressure source.

The Triservice apparatus is used in 'push-over' mode when using the compPAC ventilator. In this configuration – vaporizer between ventilator and patient – the pumping effect causes a slight-to-moderate increase in delivered vapour concentration. Capnography, end expired agent concentration monitoring and pulse oximetry should all be available at the mobile field hospital, and will enable safe anaesthesia to be provided with such equipment.

Joint operations with other national forces, which are the nature of current deployments of the British military, encourage the use of a more common platform of equipment where major medical facilities may be manned by non-UK personnel and where portability is not such a priority. More 'conventional' equipment may, therefore, be seen at a '*role 3*' hospital (*role 1* being closest to site of wounding, and *role 4* being specialist services in the UK) such as in Camp Bastion in Helmand Province, Afghanistan (Fig. 27.12).



Figure 27.12 Fabius Tiro M military configuration anaesthesia machine. An electronically controlled piston driven ventilator allows ventilation in the absence of a pressurized gas source.
Photograph courtesy of Dräger Medical, UK.

Nuclear biological chemical (NBC) capability

Under conditions of NBC warfare, mobile military operating theatres are designed, by the use of air filters, to provide a protected environment. For additional patient protection, an NBC filter may be placed on the end of the inlet tubing of the Triservice apparatus, ventilator or oxygen concentrator. These filters usually combine a glass-fibre particulate filter to intercept biological agents, chemicals in aerosol form and radioactive particles and an activated carbon membrane to adsorb toxic agents in vapour or gaseous form.

Equipment for other battlefield anaesthetic techniques

Total intravenous anaesthesia is also advocated for military anaesthesia as it can simplify equipment requirements. Ketamine/midazolam and propofol/alfentanil (either combination with vecuronium added for controlled ventilation) have been used very successfully, and requires little more than a self-inflating bag and oxygen source, together with an infusion bag (or syringe) with the mixed combination of drugs, and an infusion pump if available.[12]

ABNORMAL AMBIENT PRESSURES

Altitude

It may be necessary to use anaesthetic equipment at low ambient pressures, as in the transfer of patients by aircraft and in high-altitude locations. The highest human habitation is at about 5000 m or 16 000 ft, giving an atmospheric pressure of about 400 mmHg. Commercial aircraft, however, usually have cabin pressure maintained at 640 mmHg minimum, which is equivalent to 1500 m, despite flying at heights of over 9000 m. In order to provide safe anaesthesia, a knowledge of the altered performance of anaesthetic equipment at different ambient pressures is essential:

- *Flowmeters.* The reduction in gas density at altitude results in under-reading of variable orifice, constant differential pressure flowmeters. The error is about 20% at 3000 m. Under hyperbaric conditions, these flowmeters will over-read.[13]
- *Pressure gauges.* These are calibrated at sea-level and so over-read at altitude. The error is negligible, as the pressures measured are so much greater.
- *Vaporizers.* Saturated vapour pressure is a function of temperature, not ambient pressure. Hence, the concentration delivered by a vaporizer is

inversely proportional to the ambient pressure as the vapour pressure takes up a higher proportion of the ambient pressure at altitude, and a lesser proportion under hyperbaric conditions. However, the partial pressure of the agent, which determines the clinical effect, remains constant. Therefore, when vaporizers are used at a given setting, the anaesthetic will be delivered at a constant potency or effect, regardless of concentration changes with altitude (or depth). A vaporizer set at 1% at sea level will deliver 1.7% at 4500 m, but the clinical effect will be unaltered.

- *Gas analyzers and capnography.* Gas analyzers measure the partial pressure of the gas under test but are calibrated in percentage at sea-level. They will, therefore, under-read at high altitude. Reduction of atmospheric pressure may also affect capnography in the following ways:[14]
  - pumping of gas through sample chamber – more powerful pump may be required to maintain flow rates
  - calibration inaccuracies may occur – this may be corrected by recalibration at altitude
  - fall in barometric pressure may be electronically sensed as a gas leak within the monitor.
- *Venturi-type oxygen masks.* These will entrain less air at altitude and so deliver higher concentrations of oxygen. A 35% mask will deliver approximately 41% oxygen at 3000 m.
- *Ventilators.* Volume or time-cycled ventilators may be preferable to pressure-cycled ventilators, but capnography and other monitoring will assist in adjusting ventilator settings under these conditions.

Hyperbaric chamber and anaesthetic equipment

A brief synopsis of some issues specific to high ambient pressure environment is given below (see also Chapter 7, Oxygen delivery at high or low atmospheric pressures):

- Gas diffusion into gasses or liquids causes bubbles on decompression. Rapid decompression with gas expansion may result in breakages of sealed, and particularly glass containers.
- Oxygen-rich environment under hyperbaric conditions increases the risk of fire; normally non-flammable materials may become flammable.
- During decompression, humidity increases, causing water condensation.
- The combination of increased water vapour and metallic walls and equipment increases the risk of electrocution and short circuits.

Most anaesthetic equipment is at best untested under these conditions, and at worst dangerous. The following issues should be considered:

- Batteries – risk of bursting or leaking.
- Endotracheal tube cuffs should be liquid filled.
- Intravenous lines must be primed without air.
- Chest drains must be vented to chamber air through flutter valves (not bottles, as water seals may be problematic, particularly during rapid transition to different pressures).
- LCD monitor screens may crack or break due to gas bubbling out of solution during rapid decompression.
- Pressure transducers may malfunction.
- Simple minute volume divider ventilators may function better than other types of ventilators.
- Defibrillators are extremely hazardous due to the risks of electrocution and fire (see above). The metal floors and walls of the chamber mean that patient and operator are permanently earthed. Sparking may be disastrous under hyperbaric conditions in an atmosphere which may be contaminated with additional oxygen from the patient.

MONITORING

The minimum standards of monitoring for safe anaesthesia have been recommended by the Association of Anaesthetists. These include:

- pulse oximetry
- electrocardiogram
- arterial blood pressure
- capnography and gas analysis.

In developing countries, various local factors may make this ideal difficult to attain, in particular:

- capital cost of equipment
- reliable power source
- availability of disposables – electrodes, tranducers, etc.
- maintenance of equipment
- transportability of equipment
- user ability to interpret results.

In the absence of electronic monitoring, relatively safe anaesthesia employing minimal equipment may still be achieved with good clinical skills, using for example:

- praecordial stethoscope
- sphygmomanometer
- finger on pulse
- patient's colour (mucous membranes)
- capillary refill time
- other clinical observations.

In remote situations where monitoring may be minimal by comparison with standards of practice in the developed world, there is no doubt that both the safety and quality of anaesthesia depends heavily upon the ability of anaesthetists to appropriately adapt techniques and equipment

to the local environment and upon their skill and attention in responding rapidly to clinical signs.

ESSENTIAL EQUIPMENT TO PACK

When going to work as an anaesthetist in a developing country, there is a limit to the amount of equipment that can be taken. Excess baggage is currently charged, though some airlines will waive charges if contacted in advance and charitable status established. Communication by the now ubiquitous e-mail will determine what equipment is available. Examples of equipment to take include:

- Bodok seals;
- assorted connectors and adaptors
- basic airway equipment – self-inflating bag, face mask, Guedel airways, favourite supraglottic airways in a selection of sizes, laryngoscope (including spare batteries and bulbs)
- stethoscope
- peripheral nerve stimulator and stimulating needles, batteries
- selection of emergency drugs, relaxants and local anaesthetics, preferably in plastic ampoules (a Home Office visa is required to take controlled drugs)
- rolls of adhesive tape.

REFERENCES

1. Association of Anaesthetists of Great Britain and Ireland. *Recommendations for standards of monitoring during anaesthesia and recovery*. 4th ed. 21 Portland Place, London W1B 1PY: AAGBI; 2007. http://www.aagbi.org/publications/guidelines.htm
2. Association of Anaesthetists of Great Britain and Ireland. *Provision of anaesthetic services in magnetic resonance units*. 21 Portland Place, London W1B 1PY: AAGBI; 2002. http://www.aagbi.org/publications/guidelines.htm
3. Sweeting CJ, Thomas PW, Sanders DJ. The long Bain breathing system: an investigation into the implications of remote ventilation. *Anaesthesia* 2002;**57**: 1183–6.
4. Jackson E, Tan S, Yarwood G, Sury MRJ. Increasing the length of the expiratory limb of the Ayre's T-piece: implications for remote mechanical ventilation in infants and children. *Br J Anaesth* 1994;**73**: 154–6.
5. English WA, Tully R, Muller GD, Eltringham RJJ. The Diamedica Draw-Over Vaporizer: a comparison of a new vaporizer with the Oxford Miniature Vaporizer. *Anaesthesia* 2009;**64**:84–92.
6. Dobson M, Neighbour R. The International Standards Organisation is an obstacle to the development of appropriate anaesthetic equipment for the developing world. *IET Seminar Digests* 2008;**12213**:3.
7. Manley R. A new ventilator for developing countries and difficult situations. *World Anaesthesia Newsletter* 1991;**5**:10–1.
8. Eltringham RJ, Fan Qui W. The Glostavent – an anaesthetic machine for difficult situations. *ITACCS* 2001;**Spring/Summer**:38–40.
9. Fenton PM. Inhalation anaesthesia in developing countries: the problems and a proposed solution – 3. *Anaesthesia News* 2003;**191**:8–9. www.aagbi.org/anaesthesia_news_2003.html
10. Houghton IT. The Triservice anaesthetic apparatus. *Anaesthesia* 1981;**36**:1094–108.
11. Adley R, Evans DHC, Mahoney PF, Riley B, Rodgers CR, Shanks T. The Gulf War: anaesthetic experience at 32 Field Hospital Department of Anaesthesia and Resuscitation. *Anaesthesia* 1992;**47**:996–9.
12. Restall J, Tully AM, Ward PJ, Kidd AG. Total intravenous anaesthesia for military surgery. A technique using ketamine, midazolam and vecuronium. *Anaesthesia* 1988;**43**: 46–9.
13. McDowall DG. Anaesthesia in a pressure chamber. *Anaesthesia* 1964;**19**:321–36.
14. Pattinson K, Myers S, Gardner-Thorpe C. Problems with capnography at altitude. *Anaesthesia* 2004;**59**:69–72.

FURTHER READING

Dobson MB. *Anaesthesia at the district hospital*. 2nd ed. Geneva: World Health Organization; 2000.

Ezi-Ashi TI, Papworth DP, Nunn JF. Inhalational anaesthesia in developing countries. The problems and a proposed solution. *Anaesthesia* 1983;**38**:729–47.

Magee PT, Tooley M. *The physics, clinical measurement and equipment of anaesthetic practice*. Oxford: Oxford University Press; 2005.

McCormick BA, Eltringham RJ. Anaesthesia equipment for resource-poor environments. *Anaesthesia* 2007;**62**(Suppl 1):54–60. doi:10.1111/j.1365-2044.2007.05299.x.

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Chapter 28

The anaesthetist and the Medicines and Healthcare products Regulatory Agency

Stephen Fenlon

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The Medicines and Healthcare products Regulatory Agency (MHRA) is the government body responsible for regulating the use of all healthcare devices and medicines within the UK. For practitioners outside the UK, an understanding of the role and function of the MHRA, as a case study in medical devices regulation, will hopefully serve to inform and educate the individual as to the purpose and likely process of any local authorities.

The MHRA was formed in April 2003 by the amalgamation of the Medical Devices Agency and the Medicines Control Agency. It is an executive agency of the Department of Health and has a large body of personnel, including many health professionals, dealing with a wide remit within the health-care sector. This chapter will concentrate on the responsibilities of the MHRA as a regulator of medical devices. Regulation of clinical trials, licensing of medicines and pharmaceutical surveillance are examples of the many other areas in which anaesthetists may have intimate involvement with the MHRA.

All products, except medicines, used in health care for the diagnosis, prevention, monitoring, treatment or alleviation of illness or handicap are medical devices. There are thus about half a million items falling under the jurisdiction of the MHRA. While some, such as syringes, ventilators and tracheal tubes, are in routine use by anaesthetists, many other medical devices (e.g. artificial eyes, condoms and surgical supports) have little impact on our professional life. Unregulated manufacture and use of medical devices, much like unregulated medical practitioners, have the capacity to inflict great harm, both on patients and on those operating the device (Fig. 28.1).

As a regulatory body, the MHRA oversees manufacture and use of all medical devices. It also provides a wealth of information and advice to health professionals in the UK and beyond. As part of this process the MHRA depends on consumers and manufacturers to provide further details of how products perform in use; this information may subsequently modify device form, function and application.

A glossary of terms used in regulation of medical devices is given in Appendix 1.

STANDARDS

To ensure suitability for a given task, any device must be manufactured according to verified standards.

Health-care professionals are by no means the first group to recognize and demand the need for consistent, universal standards in the equipment and materials they work with; in other industries this process dates back many years. An essential requirement for mass distribution of goods is certifiable manufacturing specifications. Industrial standards grew less from a need for safety and more for reasons of commerce, but the potential to advance safe manufacture and use of equipment was soon exploited. During the early industrial revolution, widely used items were custom-built by many manufacturers.

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Figure 28.1 Cyclopropane. Used correctly, causes a rapid and smooth beginning to many anaesthetics; used incorrectly, can bring about a rapid and violent end. Misuse can have catastrophic consequences, and this agent is no longer available.

The chaos this caused, when trying to put different components together, led to the birth of the standard setting body. The first of these, the British Standards Institute (BSI), began work in 1901 specifying dimensions of rails and steel plate, and now encompasses a wide range of standards from heavy engineering to good employment practice. Attempts at international standardization commenced soon after in the electrotechnical field and culminated ultimately in the establishment of a new body: the International Organization for Standardization based in Geneva, which started work in 1947 (in the spirit of standardization, the abbreviation ISO is retained across all languages). The ISO will be familiar to many anaesthetists as the body that resolved the seemingly random sizes of breathing system connectors (Fig. 28.2) into the universal system we have today (Fig. 28.3).

These organizations produce the standards, but do not enforce them: in the majority of cases the application of standards is voluntary. Organizations can, however, provide independent inspection of products covered by a standard: a process termed conformity assessment. In the UK, BSI also has a conformity assessment function as well as being a standards body. Where BSI carries out such an assessment and a product meets the appropriate standards, the manufacturer may be entitled to affix the familiar



Figure 28.2 A selection of connectors from older breathing systems showing the lack of uniform dimensions. (Compare with Fig. 28.3.)

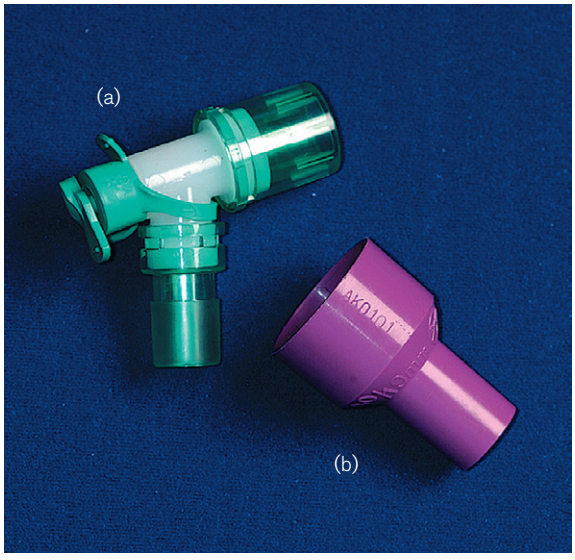


Figure 28.3 Modern 'ISO' connectors. **A.** Tracheal tube connector with 15 mm/22 mm diameter fittings. **B.** Gas scavenging connector with 30 mm diameter fitting.

BSI Kitemark to its product as a sign of manufacturing conformity (Fig. 28.4).

The ISO has also published generic standards for quality management systems. This is the ISO 9000 family of standards (ISO 9001, ISO 9002, ISO 9003). Manufacturers and many other organizations and businesses may choose, as a marker of quality, to be certified by an independent organization as meeting the requirements of the

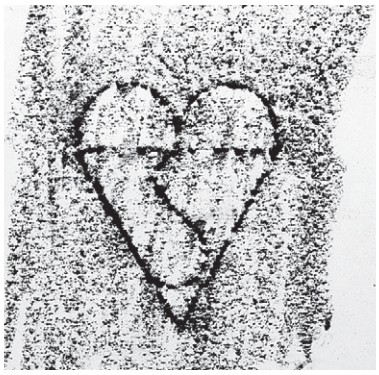


Figure 28.4 British conformity assessment: a reproduction of the BSI Kitemark taken from a drain cover.

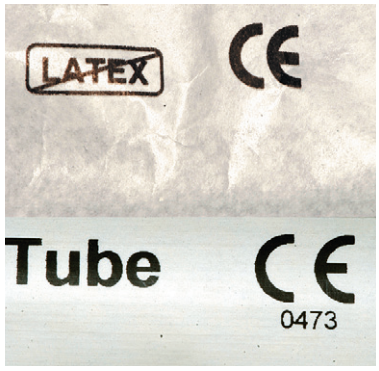


Figure 28.5 CE marks on two items of anaesthetic equipment; the lower mark also has a number with it to identify the notifying body employed for part of the conformity process.

appropriate ISO 9000 standard. This should not be confused with conformity assessment of the finished product.

Interrelationship of standards

The European Committee for Standardization (CEN, for Comitée Européen de Normalisation) can be mandated by the European Union (EU) to produce standards for the member states. This may be carried out in conjunction with ISO and national standards bodies or by wholesale endorsement of ISO standards. These are then termed European Standards and identified with the designation EN (for Norme Européenne). CEN member bodies, who are not necessarily EU members, are obliged to give such standards national status. As such, standards may often bear any or all of the prefixes BS, EN or ISO (e.g. BS EN ISO 10993–1:2009 for Biological evaluation of medical devices – Part 1).

CE MARKING

As the EU expands and integrates, there is a need for harmonized conformity assessment of the many goods passing across the borders of member states. Items deemed to require this are described in a number of directives; all medical devices are encompassed within three directives. Within each directive, essential requirements specify the standards for each item. A manufacturer confirms compliance with the essential requirements of a directive by placing the CE (Conformité Européene) mark on their product, once they have followed and complied with an appropriate conformity assessment route (Fig. 28.5). There are certain exceptions from CE marking for 'custom made' and investigational devices. Apart from these exceptions, any item specified within a directive cannot be sold within the EU without a CE mark. Regardless of which member state authorizes the manufacturer to place the mark, the item can be marketed in all member states, thus marking achieves both product regulation and compliance with the EC single market. It is inherent in the regulations that products imported from outside the EU are subject to the same set of rules.

Competent authorities and notified bodies

CE marking is of great importance to the health-care sector. Following the incorporation of the Medical Devices Directives, it is now a legal requirement for all medical devices purchased for use in the UK to have a CE mark. A major role for the MHRA is to oversee this process; it is, therefore, termed the *Competent Authority* for the UK, with other European countries having similar organizations. Exchange of information between Competent Authorities ensures a common approach throughout Europe. Hence, in essence, the MHRA is the UK arm of a large pan-European organization responsible for regulating medical devices. The directives and the standards within are produced by mutual agreement of EU member states, possibly incorporating standards from organizations such as the BSI or ISO (see above). The manufacturer is ultimately responsible for placing the mark and verifying that their product meets essential requirements. For simple low-risk items, the manufacturer can directly place the CE mark. More complex devices, such as pacemakers, require detailed independent verification and certification by an independent organization, termed a *Notified Body*, who then issue certification to allow the placing of the CE mark. A flow chart illustrates how the system works and where the MHRA may be involved (Fig. 28.6).

The implications of this whole process, for the anaesthetist in particular, and for those involved in producing and modifying medical devices, are discussed in greater detail elsewhere.[1] Although recently passed into UK law, it

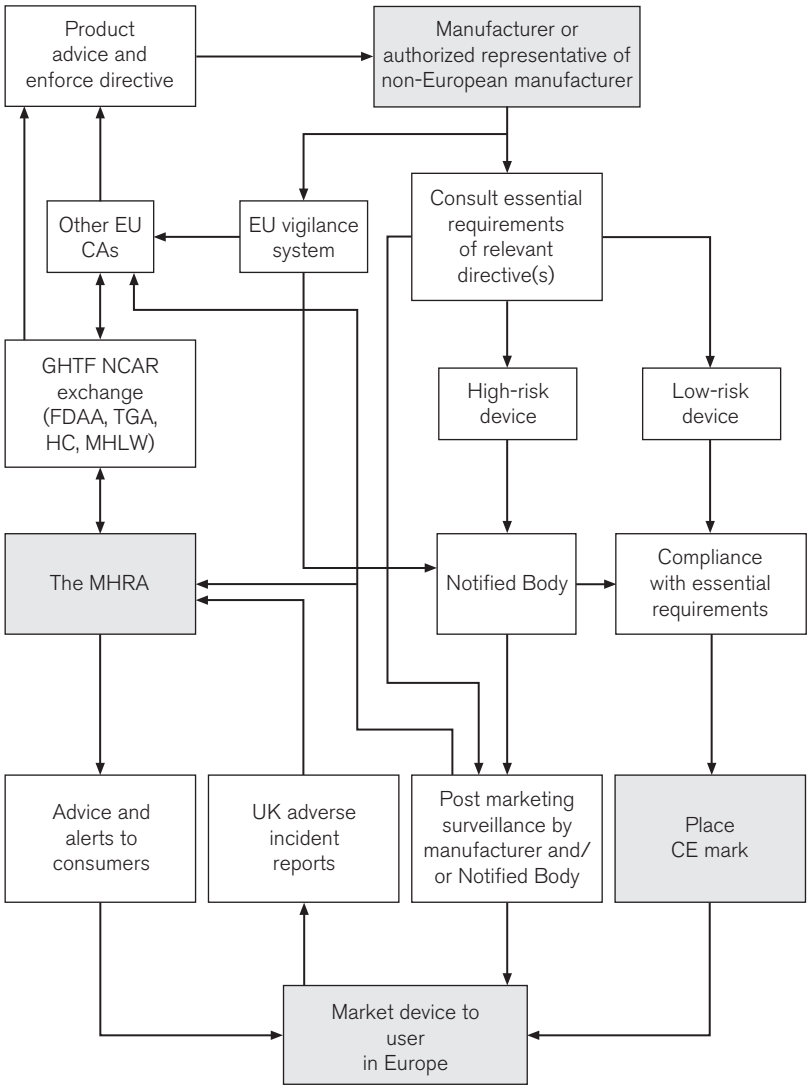


Figure 28.6 The process of placing a CE mark on a medical device, and the interrelationships of various agencies (EU, European Union; CAs, Competent Authorities; GHTF, Global Harmonization Task Force; NCAR, National Competent Authority Report, a term coined by GHTF for international reports; FDA, Food and Drug Administration of the USA; TGA, Therapeutic Goods Administration, Australia; HC, Health Canada; MHLW, Ministry for Health Labour and Welfare, Japan).

may only be a short time before a CE mark is detectable on every piece of medical equipment in UK hospitals. CE marks are of course not limited to medical devices; other directives stipulate a wide range of marketed products from detonators to jet skis, though not pedalos!

Limitations of CE marking

To the end user, the CE mark implies that the product is made to a Europe-wide standard; is compliant with the regulations; its manufacturer can be traced and is liable for the product within the terms of its prescribed use, and in correct use the benefits should far outweigh any risks. Medical devices placed on the market in the UK are regulated by the MHRA. It will investigate whether essential requirements are met, that UK Notified Bodies are suitably qualified, and, rarely, it may take action against manufacturers who fail to comply with the directive. Despite offering this reassurance, the Medical Devices Directive is not perfect and modification is needed in some areas. Devices may need to be re-classified to a higher-risk group needing more intensive pre-marketing

cannot be explored further with the reporter. Ideally, all medical device-related adverse incidents should be reported to the MHRA primarily and then additionally to other authorities as and when it is thought necessary: it may be insufficient to depend on channels of communication between organizations after the report is received. Electronic reporting by all parties is encouraged; the Manufacturers' On-line Reporting Environment (MORE) enables rapid communication of alerts from companies. Adverse incident reports are investigated according to the level of risk and incident frequency. The investigation may lead to design changes and/or improved advice to users.

SAFETY ADVICE

Information is disseminated in a number of formats to every UK hospital via its liaison officer. From 2003, the previous Hazard Notice, Device Alert, Safety Notice and Pacemaker Technical Note were replaced by a single format of notice: the Medical Device Alert. Details of the device concerned, the problem, who to distribute to, who must take action and what action to take are all included. Each alert is assigned one or more of the following categories:

- Immediate action
- Action
- Update
- Information request.

Since 2004, a new system, the Safety Alert Broadcast System (SABS), sends out alerts by e-mail and requires a liaison officer to send confirmation of receipt and subsequent action. SABS may also contain alerts from the NPSA, NHS Estates and patient safety specific guidance from the Department of Health (Fig. 28.7). The advice contained within safety alerts is not enforceable, but should be viewed as reflecting best practice. As such, a decision not to heed all or part of a warning may require a robust defence.

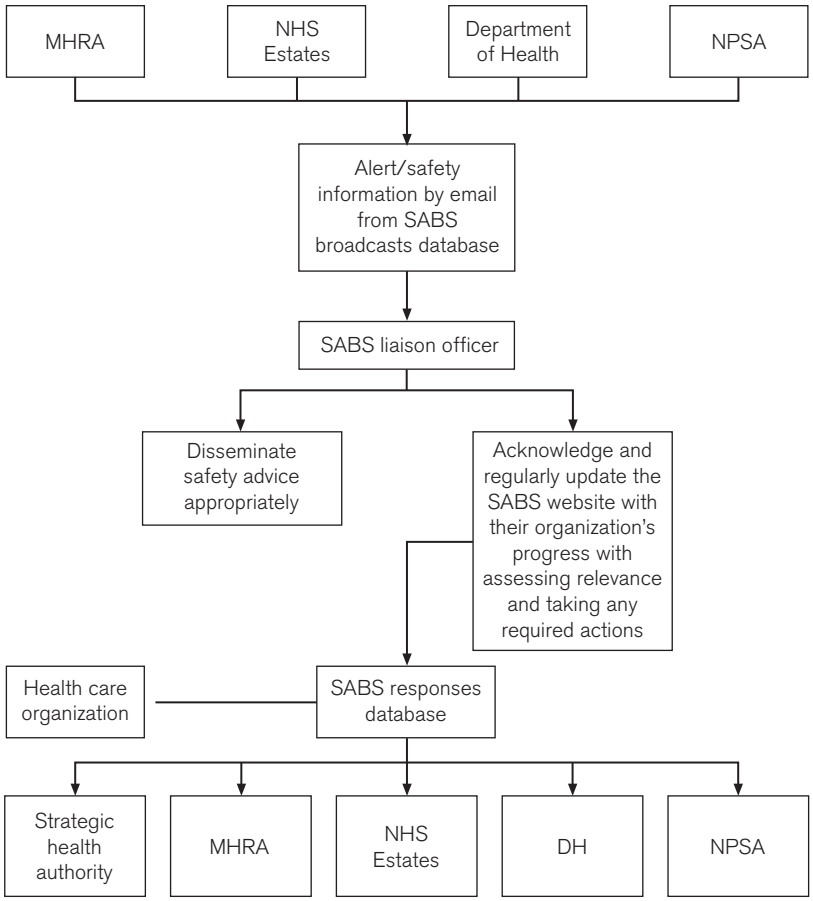


Figure 28.7 The function of SABS, the Safety Alert Broadcast System. DH, Department of Health. NPSA, National Patient Safety Agency.

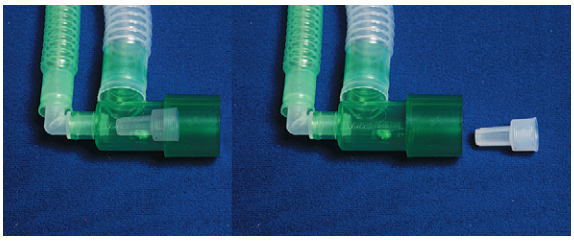


Figure 28.8 A cause of equipment failure. Occlusion of anaesthetic breathing systems with extraneous items has been reported many times.

ACKNOWLEDGEMENTS

The author would like to thank the following people for their support: Miss Katie Taylor BA MIMI, Medical Photographer, QVH; Clive Bray BSc (Pharm) MSc MRPharmS and Tony Saint BSc MSc, both of the Medicines and Healthcare products Regulatory Agency.

Appendix 1

Glossary

Adverse Incident: An event that causes, or has the potential to cause, unexpected or unwanted effects involving the safety of patients, users or other persons.

Authorized Representative: A body within the EU appointed by a manufacturer from outside the EU to enable them to meet the requirements of conformity assessment.

BSI: The British Standards Institute, the oldest such organization in the world responsible for a wide range of standards, they are also a recognized Notified Body.

CASU: The Controls Assurance Support Unit, soon to become the Healthcare Standards Unit, this body implements controls assurance and maintains and evaluates new health care standards.

CE mark: (Conformité Européene) confirms compliance with product directive standards for the EU; the size and form in which the mark appears are closely specified in Council Decision 93/465/EC.

CEN: (Comité Européen de Normalisation) European Committee for Standardization. One of three standards organizations for the EU with CENELEC (electro-technical) and ETSI (telecommunications), they develop European standards and consider applications for standards.

CNST: The Clinical Negligence Scheme for Trusts provides guidance and incentive for trusts to improve their risk management and insures them against claims for damages arising from clinical negligence.

Committee on Safety of Devices: Set up to complement the work of the MHRA by advising ministers and overseeing strategic direction in improving medical device safety and the work of the MHRA.

Competent Authority: A national organization overseeing the Medical Devices Directive; the MHRA is the Competent Authority for the UK.

Conformity Assessment: Testing of a product or process and certification that it meets a particular set of standards, usually confirmed by placing a mark somewhere on that product.

EGBAT: The Expert Group on Blocked Anaesthetic Tubing, a UK body set up to investigate the reasons for these untoward incidents and to suggest how they may be prevented.

Essential Requirements: Those agreed standards specified within a directive that a product must meet to gain a CE mark.

GHTF: The Global Harmonization Task Force, a multinational body currently chaired by the EU, which aims to harmonize features of medical devices on a worldwide scale.

ISO: International Organization for Standardization, founded in 1947, determines a wide range of international standards.

Liaison Officer: Person within a health-care organization responsible for receiving alerts from and communicating with the MHRA, it is anticipated that most liaison officers will take on the role of SABS liaison officer.

NAO: The National Audit Office is the non-governmental auditor of public spending.

Notified Body: An institution recognized by the Competent Authority of that state for the purposes of carrying out CE conformity assessment procedures.

NPSA: The National Patient Safety Agency is a special health authority created by the Department of Health as part of its drive to produce 'an organization with memory', to report and learn from problems affecting patient safety.

Patient Safety Research Programme: Undertake to research the safety of practice in certain areas; for example, a current programme is examining the risks of reuse of single-use products.

SABS: The Safety Alert Broadcast System, an initiative by the MHRA employing electronic communication to improve the flow of information for safety alerts.

Standards: Documented voluntary agreements which establish important criteria for products, services and processes. For a standard to be European, it must be adopted by one of the European standards organizations and made public.

Appendix 2

Adverse incident reporting

Advice on reporting adverse incidents with medical devices is updated and re-issued annually. The latest version can be found on the MHRA website: *www.mhra.gov.uk*

In England and Wales:
On line: www.mhra.gov.uk
E-Mail: aic@mhra.gsi.gov.uk
Enquiries and reporting: 020 3080 7080 (answering machine message has a number for out of hours urgent reports and enquiries)
Fax: 020 3118 9814
Post to:
Adverse Incident Centre, MHRA
Floor 4
151 Buckingham Palace Road
London SW1W 9SZ
Forms for reporting specific or general problems are available on the website above.

In Scotland:
On line: www.hfs.scot.nhs.uk
E-mail nss.iric@nhs.net
Tel: 0131 275 7575
Fax: 0131 314 0722
Post to:
Incident Reporting and Investigation Centre
Health Facilities Scotland
Gyle Square
1 South Gyle Crescent
Edinburgh, EH12 9EB
Edinburgh EH5 3SH

In Northern Ireland:
E-mail niaic@dhsspsni.gov.uk
Tel: 028 9052 3868
Fax: 028 9052 3900
Post to:
Northern Ireland Adverse Incident Centre (NIAIC)
Health Estates Investment Group (HEIG)
Department of Health Social Services and Public Safety
Annex 6
Castle Buildings
Belfast BT4 3SQ

REFERENCES

1. Grant LJ. Regulations and safety in medical equipment design [editorial]. *Anaesthesia* 1998;**53**: 1–3.
2. MHRA 04005. *Breathing system filters: an assessment of 104 breathing system filters*. London: Medicines and Healthcare products Regulatory Agency; 2004.
3. Cook TM. Novel airway devices: spoilt for choice? [editorial]. *Anaesthesia* 2003;**58**:107–10.
4. Wilkes AR, Hodzovic I, Latto IP. Introducing new anaesthetic equipment into clinical practice [editorial]. *Anaesthesia* 2008;**63**: 571–5.
5. Freeman M, Cooke H. Introducing new anaesthetic equipement into clinical practice [letter]. *Anaesthesia* 2009;**64**:93.
6. Park GR. Death and its diagnosis. *Br J Anaesth* 2004;**92**:625–8.
7. Medicine and Healthcare products Regulatory Agency. *Device bulletin. Managing medical devices*. DB2006(05). London: Department of Health; 2006.
8. Medicine and Healthcare products Regulatory Agency. *Devices in practice – a guide for health and social care professionals*. London: Department of Health, 2008.
9. Association of Anaesthetists of Great Britain and Ireland. *Safe management of anaesthetic related equipment. AAGBI Safety Guideline*. London: AAGBI; 2009.
10. Carter JA. Checking anaesthetic equipment and the Expert Group on Blocked Anaesthetic Tubing (EGBAT). *Anaesthesia* 2004;**59**:105–7.
11. Medical Device Agency. *Medical Device Alert: GE Healthcare Anaesthetic CareStations* (MDA/2009/026). London: Medical Device Agency; 2009.
12. Medicine and Healthcare products Regulatory Agency. *Device bulletin. Adverse Incident Reports 2008*. DB2009(02). London: Department of Health; 2009.
13. Chawla AV, Newton NI. Machine and monitoring failure from electrical overloading. *Anaesthesia* 2002;**57**:1134–5.

FURTHER READING

Bridgeland IA, Menon DK. Monitoring medical devices: the need for new evaluation methodology [editorial II]. *Br J Anaesth* 2001;**87**:678–81.

MHRA website: www.mhra.gov.uk

Association of Anaesthetists of Great Britain and Ireland. Safe management of anaesthetic related equipment. AAGBI Safety Guideline. London: AAGBI; 2009.

Chapter 29

Error, Man and Machine

Sally E Rampersad and Carlyle (Jai) Rampersad

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Medicine is classed as a high-reliability organization (HRO) in common with aviation, aeronautics and the nuclear and petrochemical industries. Mistakes are costly in human and financial terms.

Over 3000 years ago, when Icarus flew too close to the sun, the wax on his wings melted and he crashed. The accident was the first recorded case of pilot error. The next person to try flying kept the wings but declined the wax. This canny individual had learnt from someone else's mistake. 'To err is human' – we accept that error is, and always has been ubiquitous. Error is all around us, it is described in our literature and embedded in our lives.

Much has been written recently about the lessons that medicine can take from aviation,[1–3] but there is resistance to the idea of transferring those ideas from other industries to healthcare. The usual reason given is that patients and their diseases are too complex to adapt to checklists and standard operating procedures. However, it is precisely because of this complexity that tools from aviation such as these can be life-saving. In acknowledgement of this we already do refer to algorithms such as those from the American Heart Association or European Resuscitation Council (ERC) to guide us as to the best practice to follow in resuscitating the patient.[4,5]

A willingness to learn from other HROs can reduce mistakes. It is arguable that by standardizing what repeatedly must be done correctly, we are freed to use skill and creativity to address what is unique.

For the anaesthetist, some understanding of the approach to error management in an analogous system where safety is similarly critical, such as in the aviation industry, should both instruct and enhance awareness.

HUMAN FACTORS IN AVIATION

Today, almost 80% of aircraft accidents are due to human performance failures; and the introduction of *Human Factors* consideration into the management of *safety critical* systems has stemmed from the recognition of human error as the primary cause in a significant number of catastrophic events worldwide. In the late 1970s, a series of aircraft accidents occurred where, for the first time, the investigation established that there was nothing wrong with the aircraft and that the causal factor was *poor decision-making* and lack of *situational awareness*. The nuclear industry (Three Mile Island, Chernobyl) and the chemical industry (Piper Alpha, Bhopal) suffered similar accidents, where it became evident that the same issues, i.e. *problem solving, prioritizing, decision-making, fatigue and reduced vigilance*, were directly responsible for the disaster. It was clear, however, that the two-word verdict, 'human error', did little to provide insight into the reasons why people erred, or what the environmental and systems influences were that made the error inevitable. There followed a new approach to accident investigation that aimed to understand the

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previously under-estimated influence of the cognitive and physiological state of the individuals involved, as well as the cultural and organizational environment in which the event occurred. The objective of 'Human Factors' in aviation, as elsewhere, is to increase performance and reduce error: by understanding the personal, cognitive and organizational context in which we perform our tasks.

Human Factors brings into focus the fact that people are active participants in whatever they are doing, that they 'do' whatever makes sense at the time – based on the circumstances surrounding them. Individuals bring their own perspective, their own level of interest and their own state of wellbeing on the day. In many instances, they also have an emotional investment in the outcome in terms of professional pride. In other words, whereas it was once assumed that effective decision-making was the product of mechanistically making the correct choice in a rational, predictable manner every time, current understanding is that even the most superior decision-makers are vulnerable to the weaknesses of the systems in which they operate. Error, therefore, is a product of the context in which it occurs.

A commonly used diagram, which is useful in forming a basic understanding of the man/machine interface and human factors, is the SHEL model, first described by Edwards in 1972 and later refined by Hawkins in 1975 (Fig. 29.1). Each block represents one of the five components in the relationship, with *liveware* (human) always being in the central position. The blocks are, however, irregular in shape and must be carefully matched together in order to form a perfect fit. A mismatch highlights the potential for error.

For example:

- *Liveware–Environment* errors could be the product of extremely rushed procedures and items being missed. Long flights and excessive fatigue, or multiple, short busy flights, could also incur error as a direct product of the environment.

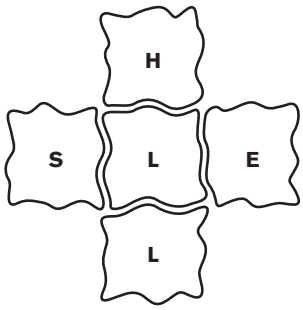


Figure 29.1 SHEL model as modified by Hawkins. **S,** software (procedures, Standard Operating Procedures Checklists, etc.); H, hardware (machine); E, environment (external, internal); L, liveware (human).

- *Liveware–Software* errors would naturally occur where information is delivered in a misleading or inaccurate manner, i.e. cluttered computer screens, or wrongly calculated weights and loads in published charts.
- *Liveware–Hardware* errors caused by poor design. Nearly 60 years ago, aircraft gear levers were located behind the pilot's seat, along with various other handles that were indistinguishable on a dark and stormy night. Mis-selection caused so many accidents that a mirror was installed on the forward panel, in an attempt to help the pilot see the levers! It was not until Boeing realized that the levers should be at the front and directly visible that these particular accidents ceased. The design was further modified by varying the shape of the individual levels, thus providing the pilot with additional tactile confirmation. An extra margin of safety had thereby been introduced into the critical function of gear selection.
- *Liveware–Liveware* errors as a direct by-product of the quality of the team relationship, interaction, leadership and information flow (see Shiva factor below).

UNDERSTANDING ERROR

Aircraft accidents are seldom due to one single catastrophic failure. Investigations invariably uncover a chain of events – a trail of errors, where each event was inexorably linked to the next and each event was a vital contributor to the outcome. The advantage in this truth is that early error detection and containment can prevent links from ever forming a chain. Errors can be defined as lapses, slips and misses, or errors of omission and errors of commission (see Appendix 1 for definition of these terms in the context of the study of error). Some errors occur as a direct result of fatigue, where maintaining vigilance becomes increasingly difficult. Conversely, others are a product of condensed time frames, where items are simply missed. Whatever its source, it must be recognized that error is forever present in both operational and non-operational life. High error rates tell a story; they are indicative of a system that either gives rise to, or fails to prevent, them. In other words, when errors are identified, it should be appreciated that they are the symptoms and not the disease.

The 'Swiss Cheese analogy' is often used to explain how a system that is full of holes (ubiquitous errors, systems failures, etc.) can appear solid for most of the time. Thankfully, it is only rarely that the holes line up or cluster (e.g. adverse environmental circumstance + error + poor equipment design + fatigue) such that catastrophic failure occurs (Fig. 29.2). Error management aims to reduce the total

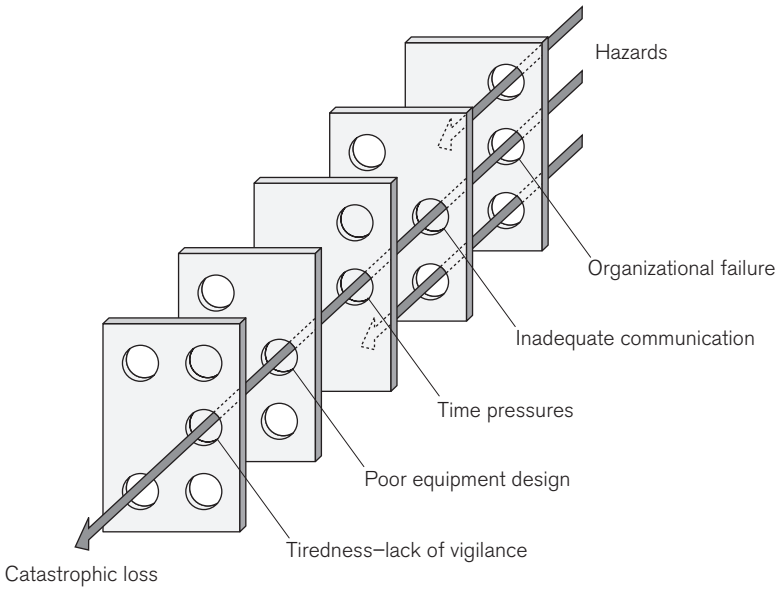


Figure 29.2 The Swiss Cheese Model, originally by James Reason,[6] demonstrates the multifactorial nature of a sample 'accident' and can be used to explain how latent conditions for an incident may lie dormant for a long time before combining with other failures (or viewed differently: breaches in successive layers of defences) to lead to catastrophe. As an example the organizational failure may be the expectation of an inadequately trained member of staff to perform a given task.

number of holes, such that the likelihood of clustering by chance is reduced (see below).

ROOT CAUSES OF ADVERSE EVENTS

Forty years ago, getting on board a commercial aircraft was a more risky proposition than it is today and arriving safely at your intended destination was not guaranteed. As the airlines focused on improving their safety record, four reasons emerged as to the root causes of these disasters:

1. Catalyst event
2. System faults
3. Loss of situational awareness
4. Human error.

The catalyst event may immediately precede the incident or seem unrelated to it: for example a decision is made to make a component part (e.g. jackscrew assembly) on an aircraft out of two dissimilar metals that will not wear at the same rate as each other (see below).

There are many examples of *system faults* that cause or contribute to errors and adverse events: scheduling systems that do not allow for adequate rest for personnel, computer systems that flag unimportant information, but fail to alert the physician about an important drug interaction.

Once the *catalyst event* has occurred and the *system faults* have contributed to, rather than mitigated the error, the patient or passengers must now rely on human beings within the system to prevent harm. The greatest tool in aviation and in medicine is *situational awareness*: the ability to keep the whole picture in focus, to not get so lost in detail that one forgets to fly the plane or to prioritize patient ventilation over achieving tracheal intubation.[7]

The final cause of an adverse event is *human error*. It is almost never the only cause and when we focus on 'who is to blame' we miss our opportunity to 'find a system to fix'.

An example from aviation

On 31 January 2000, an Alaska Airlines MD80 departed Puerto Vallarta in Mexico for Seattle, Washington, with a scheduled stop in San Francisco. Sixteen minutes into the flight the autopilot tripped off, indicating a malfunction of the autopilot or flight control system. Investigators later concluded this was due to in-flight failure of the horizontal stabilizer jackscrew and Acme nut thread assembly.

The Captain was under pressure from the airline dispatcher to return the plane to the USA, so he continued, sometimes exerting as much as 150 pounds of force on the jackscrew assembly in order to maintain control of the aircraft. What the crew did not know was that when the MD 80 was designed, dissimilar metals were used in the construction of the jackscrew assembly. Because this assembly experienced heavy flight loads, the softer

metal wore at a greater rate than the harder metal, and so the assembly became loose (**Catalyst event**).

Knowing that this could lead to catastrophic failures, the manufacturer had conducted extensive tests and issued specific procedures for inspection and lubrication of the jackscrew assembly. Other airlines, having complied with the manufacturer's approved procedures, never experienced such failures. However, because this process was difficult and time-consuming, the airline had received permission from the principal operations inspector, assigned by the Federal Aviation Authority (FAA) to that airline, to inspect at intervals of 2300 hours, instead of the 700 hours recommended by the manufacturer. When the airplane was inspected by maintenance a few days earlier, it was found to be 'outside of tolerances', but was overridden by a line supervisor, and the aircraft was placed back into service (**System faults**).

Within sight of the airport, and after recovering from a dive, the crew discussed the difficulties with maintenance. Unaware of the gravity of the situation, and ignoring the advice of the first officer who was urging him to land immediately, the captain elected to do 'a little troubleshooting' (**Loss of situational awareness and human error**). Two hours and 42 min into the flight, they lost total control of the aircraft, flying it upside down for 70 s before finally plunging into the Pacific Ocean off the coast of California.

DECISION-MAKING

Decision-makers are critically dependent on the quality of information they receive, be it verbal, computerized, or in the form of checklists or procedures. If the information is wrong, or delivered late, the decision and subsequent action will be incorrect. Decision-making is also dependent upon training, experience and situational awareness (see below).

Psychological research has shown that the human brain is able to make only a single decision at any one instant. The individual can, therefore, attend to only one process at a time and, although he can change from one process to another extremely rapidly, the danger of preoccupation is obvious. Computers (rationality) ignore the impact of time and emotion in the decision-making process. The attraction of computers to assist in collating information and decision-making is, therefore, evident.

Advancing technology has, in many instances, brought increasing complexity in the machinery, whilst at the same time seductively simplifying what is seen on screen. For instance, in older aircraft, the flight crew are surrounded by a vast number of dials, bars and indicators that provide a continuous source of raw data, which they interpret and formulate into a mental model of the aircraft's position and status. However, notwithstanding the banks of visible instrumentation, there is relatively little going on 'behind' and the logic of the systems is fairly transparent: it is quite manageable for the human brain to visualize, to interpret and to understand. Any discrepancies, either between instruments or between the instruments and the pilots' own mental model, immediately trigger investigation.

Conversely, one of the most striking features of the highly automated flight deck is the tidiness and absence of dials. The logic of the systems is now hidden deep in the computers and easy access to the 'visualization' or the 'conceptualization' of the big picture is lost. This construct is becoming increasingly pertinent in modern anaesthetic workstations with their integral monitoring and automated startup checklists. This complexity is potentially very fragile: a small problem could collapse a whole system and lure the pilot (or anaesthetist) down deceptive and unnecessary computerized pathways. Equally, complexity can also provide safeguards that we simply never see. Either way, *system action* has become invisible. Software programmes are opaque and decision-making has changed because of it. Perhaps the irony of our relationship with technology is that the more advanced the systems, the more we need the human being to solve all the problems we unwittingly designed into them in the first place.

SITUATIONAL AWARENESS

Effective decision-making is based on judgement, which in turn is based on good-quality information and accurate situational awareness. In aviation, decision-making calls for a three-dimensional appreciation of the environment (considered a hostile environment) that is travelling at speeds of up to 500 miles per hour. There is the added pressure that the consequence of error can be catastrophic. In order to minimize the potential for error, human factors training has provided flight crews with a basic understanding of the significance of mental models, perception and where and how error is most likely to occur.

The term '*situational awareness*' describes a dynamic state of (cognitive) awareness that allows for the integration of information and the use of it to anticipate changes in the current environment. Every individual has his or her own 'picture', or perception, of the environment. This mental model is informed by culture, training and previous experience. But no matter how familiar the territory, an individual's perception of a situation may not be the right one and, if left unchallenged, could lead to faulty decisions and serious consequences. Crews are, therefore, taught to confer and cross-check their mental models before making assumptions or decisions that give rise to mistakes. They are thus enhancing their situational awareness and using this knowledge to make predictive judgements on the progress of the flight (Fig 29.3).

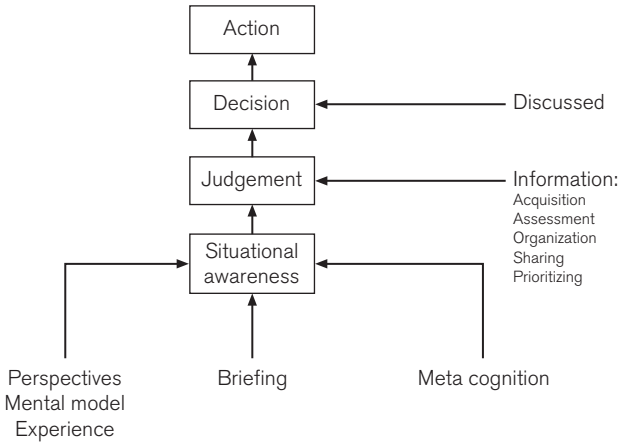


Figure 29.3 The decision-making process and factors that influence it. (Meta cognition is defined here as awareness of one's own thinking process, see text under Error Management.)

In a dynamic, fluid environment, where there is little time to go back and correct errors, it has long been appreciated that it is better to spend time making accurate decisions than to be caught in a reactive mode of trying to correct error. Errors inevitably give rise to more errors, a process that can suck its victim down a root and branch corridor of mistakes literally miles away from the original task at hand.

FATIGUE, VIGILANCE AND AROUSAL

Decision-making ability relies on vigilance so that pertinent information may be detected and it is dependent on the individual's level of arousal.

Medicine is replete with examples where *fatigue* has contributed to or caused an error. A study in interns (first year trainees) confirmed that there was a higher rate of needle stick injuries in those who were fatigued.[8] The Australian Incident Monitoring Study (AIMS) showed fatigue to be a 'factor contributing to the incident' in 2.7% of voluntary, anonymous, self-reported incidents and words such as 'haste, inattention, failure to check' often appeared in these reports.[9] Fatigue positive reports most commonly involved pharmacologic incidents such as syringe swaps and over- or under-dosage. A study of anaesthesia trainees at Stanford confirmed that chronic fatigue can be as harmful as acute fatigue.[10]

Vigilance, from the Latin *vigilantia* for watchfulness, is defined in the dictionary as 'being keenly alert to danger'. In 1943, research by the Royal Air Force showed that vigilance, requiring continuous monitoring and detection of brief, low-intensity and infrequently occurring events over long periods, is poor. This is illustrated in Fig. 29.4, which shows rapid fall-off in vigilance after a period as short as half an hour. In acknowledgement of this, modern anaesthesia machines and monitors allow alarm parameters to be set for all measured variables; this level of *vigilance monitoring* was unheard of 25 years ago (see Chapter 4, The *Anaesthetic Workstation*).

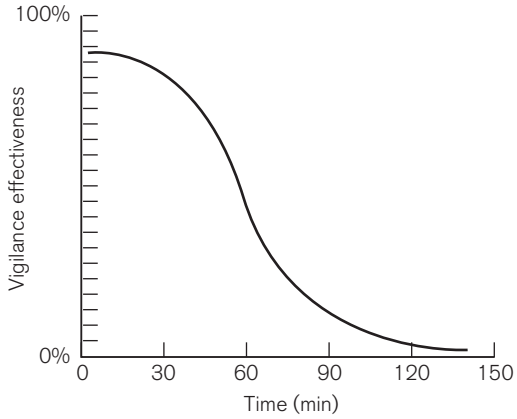


Figure 29.4 The kind of vigilance effect that can be expected in the performance of passive tasks with a low signal rate. This shows a notable decline in performance after about 30 min. This would occur over a similar period of uneventful anaesthesia.
Reproduced from Hawkins FH (1987) Human Factors in Flight, *London: Gower Technical Press.*

Arousal is the level of 'wakefulness'. For any task there is a level of arousal at which one performs most efficiently, as shown in Fig. 29.5. Surprisingly, this optimal level decreases as the difficulty of the task increases. As such, overarousal often occurs in emergency situations when

Table 29.1 Stressors contributing to reduction in performance

ENVIRONMENTAL STRESSES	DOMESTIC STRESSES	PROFESSIONAL STRESSES
Heat	Marriage	Colleagues
Noise	Divorce	Management
Vibration	Birth	Legal problems
Low humidity	Bereavement	Lack of sleep
	Financial problems	Hunger and thirst
	Health	Lack of rest

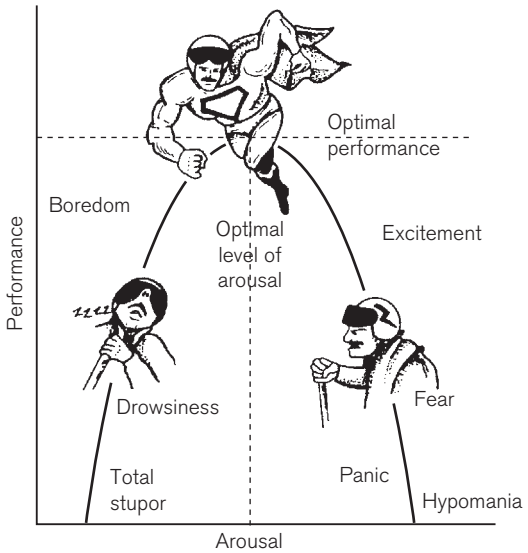


Figure 29.5 The inverted 'U' curve of arousal. Optimum performance for a particular task does *not* occur at maximum arousal. The optimum level varies depending upon the task.

difficult tasks may need to be carried out rapidly. Underarousal for a particular task slows decision-making and makes it less accurate and additionally reduces vigilance. Underarousal also occurs with boredom and sleep deprivation. Table 29.1 shows some common stressors, which adversely affect performance.

COMMUNICATION STYLES

In 2000 Sexton and others reported on 'error, stress and teamwork in medicine and aviation'.[2] They found that at that time, although most intensive care medical staff and pilots were in favour of a flat hierarchy, only 68% of surgeons favoured this. A flat hierarchy is not the answer to all of our communication difficulties in the operating room, in fact it is important that decisions are made and that at times we each need to assume the role of team leader. However, it is critical in the operating theatre that the anaesthetist is open to receiving input from others, including trainees and others whom he may consider to be in a subordinate role.

The Shiva factor

As co-authors with perspectives from both the commercial airline industry and paediatric anaesthesia, we find ourselves with the rare facility to see correlations within medicine and aviation. Here we will consider the personality types and behaviour patterns that may be error prone.

We believe that there are four identifiable steps which humans take that lead to adverse events. These we have collectively termed the 'Shiva factor', named after the head of the Hindu Trimurti, Shiva, who can be both the preserver of life, and the destroyer.[1]

The four steps are:

1. Posturing
2. Analyzing the facts to support a decision you have already made
3. Disregarding the advice or input of others
4. Persisting in a course of action despite a deteriorating condition due to loss of situational awareness.

The 'Shiva factor' in action was vividly displayed on 13 June 1982, at Washington DC National Airport. An unusually severe snowstorm had closed the airport, but by mid-afternoon it was reopened, and several aircraft had departed safely. An Air Florida Boeing 737-200 (Palm 90) then crashed in the Potomac River, hitting the 14th Street Bridge with the loss of 78 lives. The postcrash investigation revealed that the flight crew had used inappropriate de-icing procedures. The captain, possibly unaware of the proper procedures (as he was from a Florida based airline with little experience in severe winter weather; lack of **proficiency**), 'postured' in an attempt to mitigate his own fears and concerns as the crisis developed. His tone of voice on the cockpit voice recorder did not convey appropriate concern about the prevailing conditions (**Shiva factor 1**) and when his first officer questioned him, he was quick to silence him, attempting to appear as if he had everything under control (**Shiva factor 3**). In attempts to push back

from the gate, the Captain used reverse thrust to aid the tug. This was not a wise or approved procedure for the existing conditions. The snow, which was blown forward, plugged the pitot tubes on the intake of the engines. The pressures from the pitot system are used to generate an engine pressure ratio (EPR), which pilots use to confirm their power settings. The flight crew failed to turn on the engine anti-ice system, which should have prevented the excessively high false reading. All the other indications showed that the EPR gauge was wrong. The first officer brought this to the attention of the captain who has the sole authority to reject the take off. Initially he was ignored (**Shiva factor 3**). When he again questioned the accuracy of the indications and questioned the captain's assessment, he was overruled. (**Shiva factor 2**, **Shiva factor 4**). They continued the take off at a reduced power setting (because the EPR gauge falsely indicated that they had all the power that was available). Even when the stall warning system issued the appropriate warnings (no failure of **technology)** they failed to push the thrust levers fully up (failure of **proficiency and judgement)**. That simple action, taken at the right time, might have averted the disaster.

We all have seen colleagues who unfortunately have exhibited the Shiva factors in the operating theatre.

Volant diagram

This reverse Volant diagram (Fig. 29.6) illustrates a global view of a situation. Red indicates a critical situation, yellow/amber indicates a situation that is not perfect but may be good enough to allow the team to stabilize the patient and to think about how best to resolve the problem.

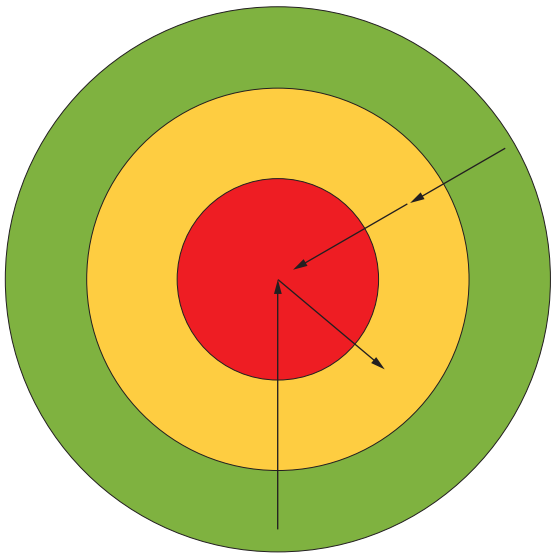


Figure 29.6 A reverse Volant diagram (see text).

Green indicates an optimum situation with all systems functioning as intended. If a catastrophic (red) situation arises our tendency is to want to go back to green quickly and to hope that no one noticed. An example of this in practice is the unanticipated difficult airway. Faced with a patient who is difficult to intubate, our tendency is to want to try to accomplish that task, especially if we did not expect it to be difficult. We want to be back in the green and we hope no one will notice (Shiva factor 1). However, repeated attempts to intubate the patient may worsen the situation resulting in the feared combination of cannot intubate/cannot ventilate. Guidelines from the American Society of Anesthesiologists and the Difficult Airway Society in the UK[11,12] do not recommend repeated intubation attempts and yet anaesthetists still have a tendency to persist in this path (Shiva factor 4). A much safer path is to return to mask ventilation or to place a laryngeal mask airway, an amber/yellow situation. This buys thinking time, so that additional **technology** may be brought in (difficult airway equipment), additional **proficiency** may be added (more/different personnel), guidelines can be referred to (**standard operating procedure**) and the **judgement** of the individual trying to intubate the patient can be enhanced by using the collective wisdom of those around him.

ERROR MANAGEMENT

Helmreich defines error management as 'the process of correcting an error before it becomes consequential to safety'.[13]

One of the principal methods of error avoidance/containment enshrined in flight safety is the use of 'briefing' and an understanding of its role in threat and error management. Every critical aspect of flight, and the conditions along the way, represents a potential threat that could cause the pilots to err. Such threats are referred to as *'red flags'*. Early identification and planning against threats reduces the likelihood of error. It also increases vigilance at those times when threats are anticipated.

Every flight begins with planning and briefing. The details of the flight plan and the reasons behind each decision are fully discussed and understood by everyone. The core purpose of the briefing is to establish a *mutual mental model* between crew members prior to departure and, equally importantly, to provide the opportunity for any additional information, relevant experience, or even subjective opinion, to be aired and added to the crews' collective situational awareness. It is recognized here that a steep authority gradient stifles information flow and a 'superior' attitude can induce stress and provoke errors in the subordinate. The preparedness consequent to adequate and appropriate planning and briefing affords the crew more mental capacity when variances occur, as they inevitably will and do.

Briefings continue throughout all major phases of flight. At particularly crucial phases of flight, i.e. approach and landing, the briefing rate increases and the 'challenge/response' use of checklists becomes more critical in error capture and mitigation. In this instance, the external environment is considered replete with potential 'threats' which, if they do not recognize and manage, will cause the crew to make errors. The greater the understanding of the threat posed by the circumstances, the less the likelihood of error arising. It is perhaps this discipline; of briefing, conferring and cross-checking, that most markedly distinguishes the aviation industry from anaesthesia and medicine.

Perhaps one of the most successful training initiatives has been the introduction, in the late 1980s, of 'real-time' flying scenarios, known as line orientated flight training (LOFT), which has enhanced the *meta cognitive* (awareness of one's own thinking process) aspect of aircrew training. These simulator scenarios are less technical and instead present the crews with situations that have many options and/or priorities of varying complexity. The 'consequences' are the product of the quality of the decisions and subsequent actions. It is virtually certain that successively poor decisions will result in a technical failure. Crews experience the outcome of their own decision pathways in a safe learning environment. The crews debrief the simulator session using human factors guidelines as performance criteria, with an instructor who is specifically accredited in this process.

BARRIERS FOR SAFETY

Just as there are four root causes of adverse events, there are also four main barriers for safety, which may trap the error or mitigate the consequences:

- technology
- proficiency
- standard operating procedures, including checklists
- judgement.

Checklists are specifically addressed below, whilst the other categories are referred to in context in the text adjoining this section.

Checklists

Checklists[14] introduce consistency and uniformity in performance, across a broad spectrum of individuals and situations, and form the basis for successful implementation of evidence-based standard operating procedures.

Human memory is notoriously unreliable for consistency in complex procedures. It worsens if you add stress and fatigue.[15] In a hierarchical situation, the mandatory use of checklists can empower subordinates to insist on the adherence to approved and safe procedures.

In aviation, checklists are used in both normal and abnormal situations. They may be done individually or in a pair, with one pilot doing and the other confirming each step. Checklists may be simply a list, or may follow a flow pattern or algorithm. For abnormal checklists, there may be a quick reference handbook, with supplemental information in a pilot handbook. A similar situation in anaesthesia would be a cardiac arrest due to local anaesthetic, where one would initially refer to the advanced cardiac life support (ACLS) or ERC guidelines, but may then refer to other materials for guidance about Intralipid dosing,[16] which is specific to this situation.

In an emergency situation, such as an engine fire, the approach was to use a memorized checklist. In medicine we rely on memorized lists, not wishing to appear lacking in knowledge. However, in highly stressful situations, items may be forgotten, so aviation has moved to a system where memorization of the first critical step is expected but then an electronic or paper checklist is used. An example would be a rapid decompression incident where the first step is that the pilot must secure his or her oxygen mask, subsequent steps are done using a checklist. The non-flying pilot reads and performs each step on the checklist, the flying pilot confirms steps, but has no other responsibility besides flying the plane. An example from anaesthesia would be the management of malignant hyperthermia. This rare complication will only be encountered by few anaesthetists during their career, but each must know how to respond. The first step is to discontinue trigger agents and then to 'turn up the oxygen and call for help': advice that has been given to trainees for years. All subsequent steps may be guided by a written or web-based checklist.[17]

When using checklists it is critical to appreciate the importance of resuming if a checklist is interrupted. Checklists should not be applied blindly without using confirmatory evidence to confirm that the action makes sense.

An example from aviation

On British Midland Flight 92, on 8 January 1989, cross wiring of the engine fire warning system gave false information when the pilots were dealing with an engine fire emergency. There were other indicators available to allow the correct course of action. While the captain was in the process of confirming which engine was on fire, by checking the corresponding gauges, he was interrupted by air traffic control (ATC) clearing him to a lower altitude. He did not resume the checklist after his transmission to ATC, but proceeded to shut down a perfectly good engine. The engine which was failing and about to disintegrate remained the sole source of propulsion. The aircraft crashed on a motorway just short of the runway with massive loss of life (Fig. 29.7).



Figure 29.7 British Midland Flight 92 crash site.

An example from anaesthesia

An anaesthetic trainee working without immediate supervision performs what he thinks is an adequate machine check from memory and fails to check the integrity of the breathing system. The reservoir bag has a large split in the wall along one fold so that it is not visible. The patient is anaesthetised and is temporarily apnoeic. The anaesthetist attempts bag and mask ventilation and the patient starts to desaturate.

The **catalyst event** is the absence of the senior anaesthetist due to illness. The **system fault** lies in allowing an inexperienced trainee to anaesthetize without adequate supervision. The outcome of this incident will now depend upon the **situational awareness** of the anaesthetist. Will he persist in his attempts to ventilate with a split bag or will he realise that he is now dealing with equipment failure?

In analyzing the event, if attention is only on the trainee, then the opportunity to 'fix the system' will be missed. If, however, the response is to impose a 'sign off' of a written equipment checklist and to prohibit inexperienced anaesthetists from working alone in remote areas, then many more critical situations will be prevented. In this scenario **technology** can not be improved upon: reservoir bags cannot be made indestructible. Lack of **proficiency** will make the outcome worse if the anaesthetist does not have the knowledge and skills to carry out the necessary emergency procedures. **Standard operating procedure** will greatly assist the team in this situation: use of a self-inflating bag in the first instance and a call for senior help. The **judgement** of the anaesthetist will be important. He must recognize that the situation has changed and he must not 'posture' but must declare the emergency and get appropriate help.

CONCLUSIONS

Anaesthetists can take many important lessons from aviation and can usefully adopt tools, such as checklists and standard operating procedures, into their practice. Communication styles such as closed loop communication and effective team dynamics are now being specifically taught in simulators and resuscitation courses.

When we analyze adverse events and attempt to learn from them, a framework such as the four causes for adverse events (catalyst event, system fault, loss of situational awareness and human error) and the four barriers for safety (technology, proficiency, standard operating procedures, judgement) can help us to look for a system to fix, not someone to blame.

Finally, if we pay attention to the Shiva factors in ourselves and others, we enhance our judgement and can lessen the 'human error' contribution to adverse events.

Happy landings!

Appendix 1

Glossary

Error of commission: Generic term for error arising from an intended act against recommended procedure.

Error of omission: Generic term for error arising from an item or action being unintentionally missed.

Lapses: Unintended 'oversight' of information after the planning stage and before action.

Mishaps: Generic term for an unfortunate event.

Misses: Errors arising from items overlooked at planning stage.

Mistakes: Error incurred when a properly selected and applied action will not result in the desired outcome.

Slips: Unintended actions/inactions occurring during the execution of familiar or pre-planned procedures (slip of memory).

Violation: A deliberate action against the recommended procedure, which may, however, be the safest choice at the time. These occur when there is a conflict between the task and the recommended procedure versus the safest outcome.

REFERENCES

1. Rampersad C, Rampersad SE. Can medicine really learn anything from aviation? Or are patients and their disease processes too complex? *Semin Anesth* 2007;**26**:158–66.
2. Sexton JB, Thomas EJ, Helmreich RL. Error, stress, and teamwork in medicine and aviation: cross-sectional surveys. *Semin Anesth* 2000;**320**:745–9.
3. Sexton JB, Marsch SC, Helmreich RL, Betzendoefer D, Kocher T, Scheidegger D, and the TOMS team. Jumpseating in the operating room. In: Henson L, Lee A, Basford A, editors. *Simulators in anesthesiology education*. New York: Plenum; 1998. p. 107–8.
4. ECC Committee, Subcommittees and Task Forces of the American Heart Association. American Heart Association guidelines for cardiopulmonary resuscitation and emergency cardiovascular care. *Circulation* 2005;**112**(24 Suppl):IV1–203. Epub 2005 Nov 28.
5. Nolan JP, Baskett PJF, editors. European Resuscitation Council guidelines for resuscitation 2005. *Resuscitation* 2005;**67**(1 Suppl): S1–S190.
6. Reason JT. *Managing the risks of organizational accidents*. Aldershot: Ashgate; 1997.
7. http://www.chfg.org/resources/07_qrt04/Anonymous_Report_Verdict_and_Corrected_Timeline_Oct_07.pdf Accessed July 17 2011.
8. Ayas NT, Barger LK, Cade BE, Hashimoto DM, Rosner B, Cronin JW, et al. Extended work duration and the risk of self-reported percutaneous injuries in interns. *Resuscitation* 2006;**296**:1055–62.
9. Morris GP, Morris RW. Anaesthesia and fatigue: an analysis of the first 10 years of the Australian Incident Monitoring Study 1987–1997. *Anaesthetic Intensive Care* 2000; **28**:300–4.
10. Howard SK, Gaba DM, Rosekind MR, Zarcone VP. The risks and implications of excessive daytime sleepiness in resident physicians. *Acad Med* 2002;**77**:1019–25.
11. American Society of Anesthesiologists Task Force on Management of the Difficult Airway. Practice guidelines for management of the difficult airway: an updated report by the American Society of Anesthesiologists Task Force on Management of the Difficult Airway. *Anesthesiology* 2003;**98**:1269–77.
12. Henderson JJ, Popat MT, Latto IP, Pearce AC. Difficult Airway Society guidelines for management of the unanticipated difficult intubation. *Anaesthesia* 2004;**59**:675–94.
13. Helmreich RL, Merritt AL. *Culture at work in aviation and medicine*. Aldershot: Ashgate; 1998.
14. Hales BM, Pronovost PJ. The checklist – a tool for error management and performance improvement. *J Crit Care* 2006; **21**:231–35.
15. Bourne LE, Yaroush RA. Stress and cognition: a cognitive psychological perspective. *NASA* 2003:1–121.
16. Picard J, Meek T. Lipid emulsion to treat overdose of local anaesthetic: the gift of the glob. *Anaesthesia* 2006;**61**:107–9 (editorial).
17. http://medical.mhaus.org Accessed 3 December 2008.
18. Ralston M, Hazinski MF, Zaritsky AL, Schexnayder SM, Kleinman ME, editors. *Pediatric advanced life support professional course guide, part 5 resuscitation team concept*. Dallas: American Heart Association; 2006.

FURTHER READING

Dekker S. *Just culture – balancing safety and accountability*. Farnham, UK: Ashgate Books; 2007.

Dekker S. *The Field Guide to Understanding Human Error*. Farnham, UK: Ashgate Books; 2010.

Dismukes K, Berman BA, Loukopoulos LD. *The limits of expertise: rethinking pilot error and the causes of airline accidents*. Farnham, UK: Ashgate Books; 2007.

Flin R, O'Connor P, Crichton M. *Safety at the sharp end – a guide to non-technical skills*. Farnham, UK: Ashgate Books; 2008.

The above courtesy of Team Resource Management, Terema Ltd UK

Gawande A. *Complications – a surgeon's notes on an imperfect science*. London: Picador; 2003.

Gladwell M. *Blink – the power of thinking without thinking*. New York: Back Bay Books; 2005.

Vincent C. *Patient safety*. Edinburgh: Elsevier; 2006.

UK portal for further information on human factors: Clinical Human Factors Group: www.chfg.org http://asrs.arc.nasa.gov/docs/cb/cb_346.pdf Accessed 3 December 2008.

Winters BD, Gurses AP, Lehmann H, Sexton JB, Rampersad CJ, Pronovost PJ. Clinical review: checklists – translating evidence into practice. *Crit Care* 2009;**13**(6):210. Available online: http://ccforum.com/content/13/6/210.

Chapter 30

Warming devices

C Mark Harper

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BACKGROUND

Inadvertent perioperative hypothermia (IPH), defined as core body temperature ≤36.0°C, is a common consequence of anaesthesia. It has a number of adverse effects, including greater intraoperative blood loss and consequent blood transfusion,[1] an increased rate of wound infection,[2] morbid cardiac events[3] and pressure sores,[4] as well as a longer stay in both recovery and hospital.[5] It also causes subjective discomfort.

Maintaining normothermia perioperatively can reduce the incidence of these adverse effects. There are a number of devices that can be used to this end. They may be devices that attempt to conserve the patient's own heat (*passive*) or devices that transfer heat from an external source to the patient (*active*). The latter may warm the patient externally or via warmed intravenous and irrigation fluids.

In 2008, the United Kingdom's National Institute for Clinical Excellence (NICE) produced the 'Management of inadvertent perioperative hypothermia in adults' guideline.[6] Its strength is in the fact that it is a clear endorsement of the clinical and cost benefits of perioperative warming, but its weakness is that it does not cover the full range of technology available due to a sparse research base.

The recommendations can be succinctly summarized as:

- all patients having operations lasting more than half-an-hour, or
- at high-risk of perioperative hypothermia

should receive warming. In addition, all fluid infusions of 500 ml or more should be warmed.[7,8] Together, these recommendations, therefore, encompass the majority of operations and most intravenous infusions, and highlight the need for a wide knowledge of the available warming technologies.

Temperature monitoring is covered elsewhere in this book. However, it should not be forgotten that this is an integral part of perioperative thermal management. Unfortunately, there are limitations to all currently available methods of perioperative temperature monitoring and it should be remembered that accuracy in the laboratory does not necessarily imply accuracy in the clinical setting.

PHYSICAL PRINCIPLES

These are important in understanding how warming devices work, how heat is lost and gained by the body, how warming devices work and, consequently, the best way to go about maintaining normothermia.

Heat generation

In the human body, the generation of energy is by chemical reaction and its quantity determined by the substrates and products of the reaction. Combustion of glucose and protein produces 4.1 kcal/kg, whereas fat provides 9.3 kcal/kg. Although heat generated in this way

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depends on the level of activity/metabolism, which is reduced under anaesthesia by 15–40%, most core hypothermia is the result of altered distribution of body heat rather than alterations in the balance of heat production and dissipation.

Heat transfer

Heat transfer can only take place down a temperature gradient. Within the body there is a gradient between core and peripheral 'compartments'. Peripheral tissues are usually 2– 4°C cooler than the core. There is then the much more variable gradient between the peripheral tissues and the environment. This simplistic model is, however, somewhat modified by the body's control over heat distribution via the circulation. The importance of this is demonstrated by the fact that even during warming, the peripheral compartment remains about 1°C less than the core temperature.

Conduction

This is defined as the transmission or conveying of energy through a medium without perceptible motion of the medium itself. In terms of heat, this is the transfer of thermal energy through a substance from a higher- to a lower-temperature region. Heat conduction occurs by atomic or molecular interaction. Core cooling occurs through conduction loss to the cooler peripheral tissues. In the same way, temperature loss through the infusion of cold intravenous fluids may, for practical purposes, be considered a conductive loss.

Convection

This is defined as the transfer of heat through a fluid medium (liquid or gas) caused by molecular motion. This is an important route of heat loss, wherein heating of air in contact with the body causes it to expand and rise resulting in the formation of convective currents that carry away heat. These losses are greatly exaggerated wherever external air currents remove this warmed air more rapidly. It is is the dominant source of heat loss in environments such as laminar flow operating theatres.

Radiation

This is defined as the transfer of heat from one surface to another via photons. It is not, therefore, dependent on the temperature of the intervening air. It is dependent upon the emissivity of two surfaces and the difference between the fourth power of their temperatures in degrees Kelvin. A black body absorbs and emits heat perfectly and has an emissivity of one (the opposite is a perfect mirror). Human skin acts more like a black body having an emissivity of 0.95 for infrared light. This is a significant source of perioperative heat loss.

Evaporation

This is the process whereby heat energy is used to change water from a liquid state into a vapour. This phenomenon which causes an observed heat loss, is governed by the latent heat of vaporization of water and consumes 0.58 kcal for each gram of sweat evaporated. Under normal circumstances and in the absence of sweating evaporation contributes only 10% of heat loss, mainly due to losses from the respiratory tract. However, its contribution becomes quite substantial during operations in which there is significant visceral exposure. It has also been suggested that alcoholic skin preparations may decrease the temperature of a 70 kg person by up to 0.7°C.

Thermal capacity

This is defined as the amount of heat energy required to increase the temperature of a unit quantity of a substance by a specific temperature interval and is significant in both the loss of heat and its prevention. All alterations in body temperature are the result of changes in the heat content of the tissues.

The specific heat capacity of body tissue is 0.83 kcal/kg/°C. In terms of perioperative thermal balance, of additional importance are the specific heat capacities of (dry) air, which is 0.24 kcal/kg/°C, and water, which is 1.0 kcal/kg/°C.

Insulation

Within the body this affects the conductive component of heat loss insofar as fat insulates almost three times as well as muscle. It is also the principle behind passive means of preventing IPH.

DEVICES USED TO PREVENT PERIOPERATIVE HYPOTHERMIA

Passive devices

Although ordinary blankets, bedding and clothes prevent heat loss to some extent, they are not appropriate in the setting of the operating theatre where higher standards of cleanliness are required. The first products specifically designed for this setting were called 'space' blankets. These are made from a lightweight non-permeable material incorporating a reflective layer that reduces the patient's radiant heat losses. The non-permeable element provides insulation from the operating theatre environment and

reduces the convective heat losses. Their effectiveness is partly based on the high emissivity of heat from the human body. They also have the advantage that they meet the safety standards of 'Flammable Fabrics' Acts. However, for the majority of procedures, insulation alone is insufficient in preventing heat losses during anaesthesia, surgical preparation and subsequent surgery. There is, therefore, the need to provide heat from an external source.

Active devices

Circulating water devices

Initially, prior to the advent of forced-air warming, patients were placed on circulating hot water mattresses in an attempt to counteract heat loss and maintain normothermia. In theory the high specific heat capacity of water in the mattress should be very effective at providing heat. In practice, however, these devices only effectively deliver heat to those areas in direct contact with the mattress, which constitutes a relatively small proportion of the body. Furthermore, those small areas in direct contact are under pressure and so have a compromised blood supply that reduces the amount of heat transfer even further. Additionally, in this situation the relatively high thermal capacity of the water is a disadvantage as it increases the likelihood of thermal damage, which has been described at settings of 39 °C.

Newer devices overcome these problems by circulating the water through special garments or pads. They include the Kimberly-Clark Patient Warming System (Figs 30.1A and B), which uses adhesive 'energy transfer' pads with micro-channels for circulating water that can be applied to the back, thighs, chest, or any combination of the three, depending on the site of surgery. Another modern system is the Allon circulating-water garment. This conductive heating garment is divided into separate segments for arms and thighs, which allows clinicians to cover different body surfaces depending on the site of surgery. Perhaps unsurprisingly, given the different thermal characteristics of water and air, both the above have both been shown to be more efficient at warming volunteers than forced-air devices (see below).

Carbon fibre and polymer devices

Carbon fibre heating mattresses consist of electrically conductive bundles of this material that criss-cross the device in much the same way as the wire element in electric blankets. When an electric current is passed through the device the resistance of the material causes the mattress to heat up. However, the biggest problem with these is that it is difficult for such systems to deliver uniform heating characteristics, with the consequent risk of burns to a patient. This is because the area of heating surface may be inadequate and the hardness of the bundles means that



Figure 30.1 Kimberly-Clark Patient Warming System. **A**. back pad with warm water microcirculation channels. **B**. warm water delivery and control unit.
Images courtesy of Kimberley-Clark Healthcare.

these require some form of pressure relief material on top which attenuates the warming performance.

In contrast to carbon fibre, *carbon polymer* theoretically benefits from a higher thermal transfer capacity, more uniform heating characteristics and better elasticity (see below).

The heat generated in the polymer is caused by excitation of the carbon atoms within the polymer due to the passage of an electric current. This is produced by a low-voltage source applied across the edges of the sheet. The polymer increases the electrical resistance by holding the pattern of the carbon particles. The variation in temperature across a sheet the size of an operating table is less than 1°C, thereby delivering heat in a uniform manner over a large surface area. The properties of the polymer also allow a viscoelastic foam to be placed under the warming surface which provides pressure relief superior to a standard operating table mattress or gel pad. The moulding of the foam improves the efficiency of the mattress as the heat is transferred through conduction rather than convection. It also prevents one of the problems with other warming mattresses in that there is less pressure to occlude the skin circulation, which in turn reduces the incidence of thermal damage and pressure sores.

A full-length mattress takes approximately 65W at full power (i.e. during warming up phase). The power needed to maintain temperature varies depending on patient characteristics and ambient conditions. A thermistor on the rear face of the polymer sheet is connected to a microprocessor control unit that regulates the power to the mattress to maintain the selected temperature. This can usually be set at between 37 and 40°C.

The working components are encapsulated in a latex-free cover, with welded seams, which means that the mattress can be cleaned in the same way as an operating table (Fig. 30.2).



Figure 30.2 Inditherm carbon polymer warming mattress.
Images courtesy of Inditherm Medical.

The logistical advantages of carbon technologies include that the warming area can be maintained largely irrespective of the surgical access required and the operating theatre is quieter due to the absence of circulating air and complaints from the surgical staff that they are too warm. On the other hand, in circumstances where there is reduced patient contact with the mattress (e.g. lithotomy position) equivalence with the variety of shaped ('specialist') forced-air warming blankets has yet to be established. Recent advances incorporating carbon polymer into blankets may serve to overcome problems of inadequate mattress contact area.

Forced-air warming blankets

These have revolutionized patient warming. Broadly speaking a large volume of air is blown over a 450–1400 W electrical element which warms it to 35–46°C. This is then passed through a 'quilted' blanket that covers the patient (Fig. 30.3). The power consumption is around 850 W for the lower powered devices and up to 1500 W for the more powerful ones (i.e. a factor of 10–20 greater than the carbon polymer mattresses). There is a significant variability in the performance of the different types of forced air heating devices (Table 30.1).

Various different blankets have been developed in order to maximize the surface area covered during different surgical procedures and exposures; including now forced warm air mattresses for positioning underneath the patient. With improving technology, the heating devices themselves can be much smaller and so it has been possible to develop special jackets with portable heaters



Figure 30.3 Bair Hugger forced air warming blanket.
Image courtesy of Arizant UK.

Table 30.1 Efficacy of forced air warming systems with full body blankets in a laboratory simulation. Heat exchange coefficients (h), mean temperature gradients at a calculated surface temperature of 32°C (ΔT at 32°C), 34°C (ΔT at 34°C), 36°C (ΔT at 36°C) and 38°C (ΔT at 38°C) and the resulting heat exchange between the full body blanket and the manikin

SYSTEM	H	Δ (°C) AT				HEAT EXCHANGE (W) AT			
	(WM^{-2} $°C^{-2}$)	32 °C	34°C	36°C	38°C	32 °C	34°C	36°C	38°C
Bair Hugger 505 and full body blanket	21.9	1.40	0.91	0.43	–0.06	30.7	19.9	11.4	–1.6
Bair Hugger 750 and full body blanket	28.0	2.76	2.09	1.31	0.53	77.3	58.5	44.4	18.0
Life Air 1000 S and full body blanket	26.4	1.76	1.17	0.58	–0.02	46.5	30.9	18.5	–0.6
Snuggle Warm SW-3000 and full body blanket	32.2	1.93	1.42	0.91	0.40	62.1	45.7	35.5	15.6
Thermacare and full body blanket	23.6	1.97	1.40	0.83	0.26	46.5	33.0	23.70	7.4
Thermacare and Optisan full body blanket	17.1	2.79	2.00	1.22	0.43	47.7	34.2	25.2	8.9
WarmAir and full body FiltredFlo blanket	13.4	2.61	1.83	1.05	0.27	35.00	24.5	17.0	4.4
Warm-Gard and full body blanket	15.4	3.18	2.65	2.12	1.59	49.0	40.8	39.5	29.6
Warm-Gard and reusable full body blanket	15.3	2.50	1.83	1.16	0.49	38.3	28.0	21.5	9.1
WarmTouch and full body blanket	28.1	1.24	0.74	0.24	–0.26	34.8	20.8	8.2	–8.8
WarmTouch and MultiCover full body blanket	14.5	3.18	2.46	1.74	1.02	46.1	35.7	30.5	17.9

(From Brauer A, English MJ, Steinmetz N, et al. Efficacy of forced-air warming systems with full body blankets. *Can J Anesth* 2007;**54**:34–41, with permission.)

that can be used to keep patients warm throughout the perioperative period.

There are both disposable and reusable products. What is gained in terms of reduced consumables with the latter may be partly lost by the environmental and financial costs of laundering.

Of the single-use types, there are two versions which differ in that one is a closed system whereas the other forces air out through small holes on the side of the blanket facing the patient. There is the unproven possibility that the latter may introduce pathogens into the surgical field. They both have filters on the air intake; although these are not small enough to exclude all pathogens that may also exist on and within the warming unit, there is nothing in the NICE literature review to suggest an increase in infection rates.

Radiant heaters

Radiant heaters are electric heaters that generate heat using infrared radiation in the same way that the sun heats the Earth (Fig. 30.4). The infrared spectrum has a wavelength of 0.7–10 μm. Non-industrial heaters use the medium part of the spectrum (approx. 1.5–5.6 μm), typically utilizing the range 2–4 μm. Radiant heat transfer, unlike conduction and convection, requires no intermediate conductor or convector, as infrared energy, like light, passes directly from the source to the receiver. The rate of heat transfer depends on the emissivity of the source, the absorptivity of the receiver, the difference between their absolute temperatures (raised to the fourth power), and the distance between them.



Figure 30.4 An infrared radiant warmer for infants.

The element on a typical neonatal unit typically consumes 600 W power. They can either generate heat to a set air temperature or, via a feedback mechanism, to a set skin temperature. The danger with the latter is that if the sensor falls off it will read air temperature, which will run the risk of overheating of the patient.

Radiant heaters have the advantage that they provide efficient, uniform and immediate heat, and they are unaffected by air currents, such as those in laminar flow operating theatres. In addition, they do not generate air-currents, which might facilitate the spread of pathogens.

The disadvantages are that they warm the operating staff even more than forced-air warmers, that they may be impractical for patients much larger than 8 kg and that they will only warm exposed surfaces (which can increase evaporative losses). Thus, their use is largely restricted to paediatrics.

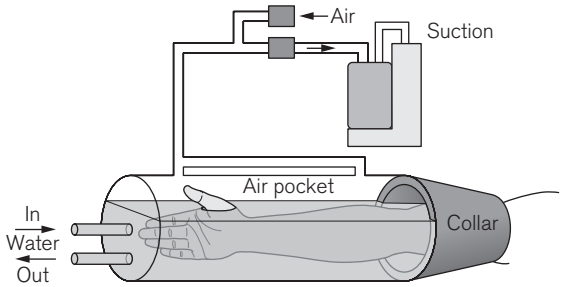


Figure 30.5 Cylindrical transparent plexiglass cylinder.
Redrawn with permission from Rein EB, Filtvedt M, Walloe L, Raeder JC. Hypothermia during laparotomy can be prevented by locally applied warm water and pulsating negative pressure. Br J Anaesth *2007 Mar;98(3):331–6.*

Other devices

A number of alternative surface warming devices have also been developed but which have not yet entered the mainstream of clinical practice. One such device is described below.

Locally applied warm water and pulsating negative pressure

With appropriate methodology it is feasible to warm the whole patient with very localized heat application. One study has shown the device illustrated in Fig. 30.5 to be more effective than forced-air warming used during laparotomy[10].

It consists of a custom-built, tube-shaped, transparent Plexiglass chamber, which is sealed to the proximal part of the arm by a neoprene collar. Warm water at 42.58 °C is circulated between the cylinder and a thermostat-regulated water bath, which is circulated at 3.5 l min^{-1}. Prior to commencing warming, the chamber is three-quarters filled, leaving an air pocket from which the air could be evacuated to give negative pressure, which is pulsated between 0 and 240 mmHg.

Devices used to warm intravenous fluids

NICE (2008, see above) in the UK has identified the risk of infusion of cold intravenous fluids as a potential cause of IPH and has issued guidance on the warming of volumes of 500 ml or more in adults. Extrapolating from the thermal capacities of water and body tissues, the infusion of 1 L of fluid intravenously at room temperature will theoretically lead to a decrease in core temperature of >0.2 °C in a 70 kg person.

Fluid warmers can be broadly classified into forced air, metal plate, circulating water and infrared devices. There are also high-flow and low-flow versions with large variations in performance especially at higher flow rates (Tables 30.2 and 30.3).

Table 30.2 Simulated clinical evaluation of four standard fluid warming devices to measure output temperature and flow rates (mean and range)*

	NO WARMING DEVICE	BAIRE HUGGER	HOTLINE	STANDARD RANGER	FLUIDO
Output temperature, °C	22.9	26.3	31.1	35.2	35.3
Flow					
Measured (ml min^{-1})	185 (184–185)	68 (68–73)	152 (152–154)	153(152–154)	141 (140–142)
Recommended**	–	≤17	≤83	≤150	15–800
% change in flow from no warmer	–	–63	–16	–16	–23

*When subject to gravity flow with 1 m head of fluid compared to no warming device (all through a standard IV set and 14 G cannula).
**Manufacturer's recommended maximum flow rates are given for delivery of fluid at 37°C.[11]
(From Turner M, Hodzovic I, Mapleson WW. Simulated clinical evaluation of four fluid warming devices. *Anaesthesia*. 2006;**61**:571–5, with permission.)

Table 30.3 Simulated clinical evaluation of four standard fluid warming devices to measure output temperature and flow rates (mean and range)*

	GRAVITY FLOW	PRESSURE-DRIVEN FLOW			
	NO WARMING DEVICE	BAIRE HUGGER	HOTLINE	STANDARD RANGER	FLUIDO
Output temperature, °C	22.9	23.9	26.3	31.2	35.5
Flow (ml min^{-1})	185 (184–185)	225 (224–228)	375 (372–380)	407(404–410)	446 (442–448)
% change in flow from gravity flow with no warmer	–	+22	+103	+120	+141

*With giving set pressurised to 300 mmHg compared to no warming device with gravity flow (all through a standard IV set and 14 G cannula).[11]
(From Turner M, Hodzovic I, Mapleson WW. Simulated clinical evaluation of four fluid warming devices. *Anaesthesia*. 2006;**61**:571–5, with permission.)

Forced-air/coil warmers

As can be seen from the entry for the Baire Hugger (Fig. 30.6) device in Table 30.2, these are the least effective and consist simply of a coil placed inside the hose of a forced air warming mattress. Their poor performance can be explained by the different thermal capacities of air and water (see above). In this case air, which has a low capacity, is being used to heat fluid which has a high capacity.

Plate warmers

In these devices the fluid passes through a special cartridge that brings it into indirect contact with an electrically heated metal plate. The temperature of the plate is set to 40°C (rather than 37°C) to compensate for the loss of heat over the length of line between the warmer and the patient. Small warmer units are also available that can be placed close to the patient's infusion site (Fig. 30.7). The water channel of the cartridge is made from plastic with one side bonded to an aluminium-serrated plate. The latter is placed in direct contact with the heater element and is responsible for the transfer of heat to fluid passing through the plastic channels in the cartridge.

Counter-current warmers

These attempts to offset the losses between a warming device and the patient by placing a circulating warm water jacket around the intravenous line along its whole length. Although the warmed water does not come into direct contact with intravenous fluids, bacteria can proliferate in the reservoir and so the possibility of cross-infection remains.

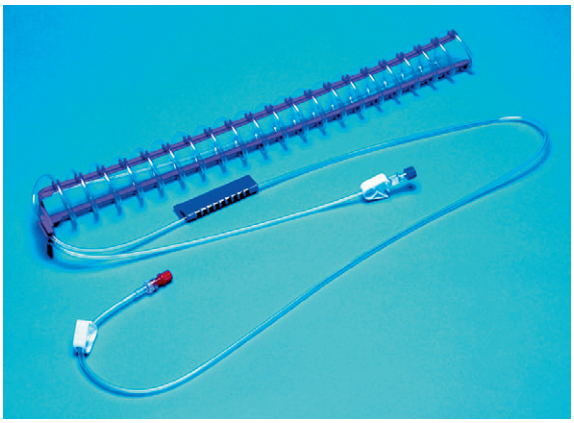
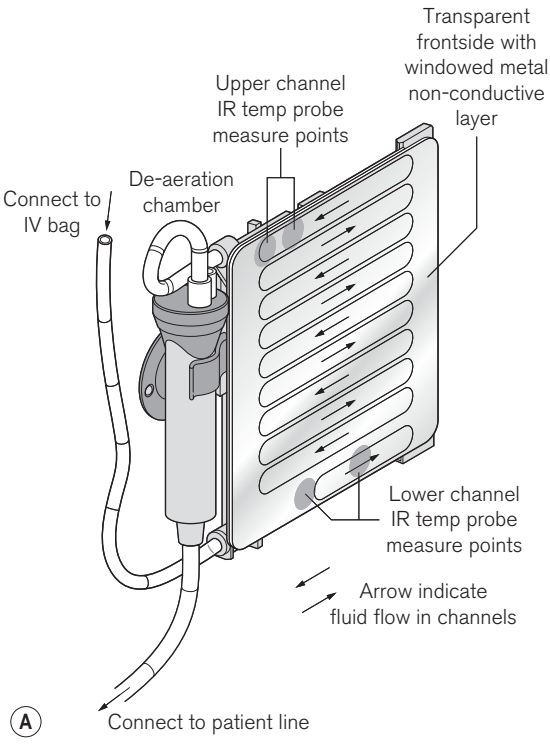


Figure 30.6 Dry air fluid warming coil.



Figure 30.7 The Vital Signs enFlow dry plate warmer. The small IV casette is shown inserted into the warming plate.
Image courtesy of GE Healthcare.



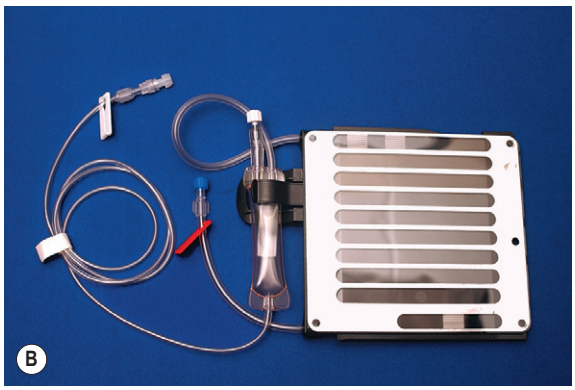




Figure 30.8 Fluido dry fluid warmer. **A.** Stucture. **B.** Insert. **C.** External appearance.
Image of Fluido device (C) courtesy of TSCI International, BV.

Infrared flow compensated fluid warmers

The Fluido system (Fig. 30.8) is a dry fluid warming system. Fluid channelled through a rigid cassette is heated by four infrared lamps with a maximum output of 1200 W.

By calculating fluid flow rate from the temperature change across two set points the device is able to alter the output temperature of the fluid to compensate for the expected heat loss along the fluid administration set to the patient. Flow rates up to 800 ml min^{-1} are claimed at temperatures of up to 39 °C at the patient.

High-flow fluid warmers

Devices designed for infusing fluids at high flow rates require both lower resistance 'cartridges' and some form of pressurization.

The 'Level One' (Fig. 30.9) system from Smiths Medical uses heated water at 42 °C and an aluminium counter current heat exchanger to deliver flow rates up to a claimed 1400 ml min^{-1}. Two rigid housings for the bags of infusate are pressurised to 300 mmHg by an in built compressor to deliver an uninterrupted flow. The maximum effective flow rate for fluids at 10 °C is approximately half of that for fluids stored at 20 °C. These devices have been inadvertently charged with bags partly containing air and have delivered fatal air emboli into patients. Although newer versions have an air detector, this can be bypassed. There is also an upgrade available for the older machines, but the risk of air embolus remains significant with any system that uses a pressurised infusion bag.

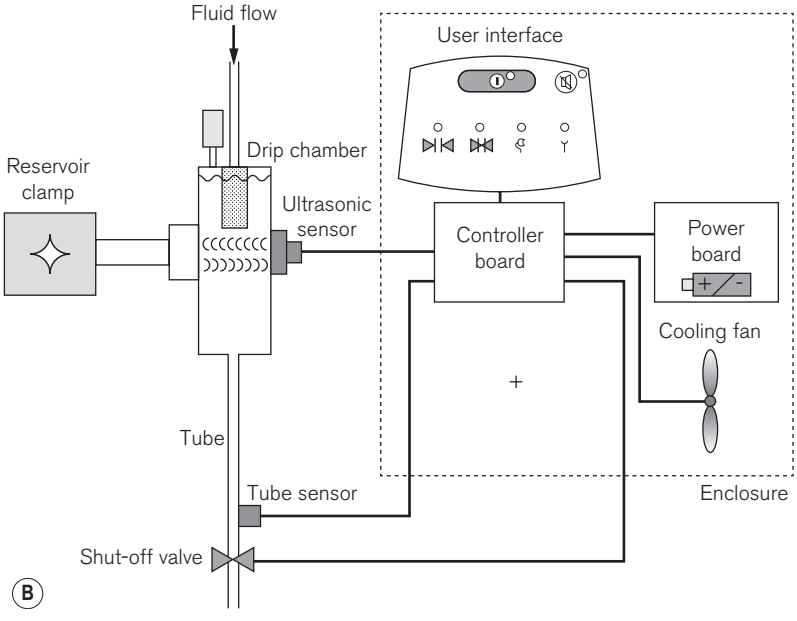
The Fluido infra red warmer (see above) can also be used as a high flow device. The casette can be fitted with two infusion lines that are inserted into IV fluid bags pressurised by pneumatic chambers fed by a compressor. The AirGuard component (Fig. 30.10) protects against air embolism. If the fluid in the chamber falls below a fixed level a valve will close the supply tubing, to prevent the infusion of air. In addition, an infrared tube sensor continuously monitors the presence of this tubing to ensure that it is fixed correctly into the shut-off valve. The complete system is mounted on a drip stand (Fig. 30.11).



Figure 30.9 Level 1, high flow fluid warming system. *Image courtesy of Smiths Medical, UK.*



A



B

Figure 30.10 (**A**) The Fluido AirGuard. (**B**) A schematic of the Fluido AirGuard system.
Images courtesy of TSCI International BV.



Figure 30.11 Fluido High flow system.
Image courtesy of TSCI International BV.

REFERENCES

1. Rajagopalan S, Mascha E, Na J, Sessler DI. The Effects of Mild Perioperative Hypothermia on Blood Loss and Transfusion Requirement. *Anesthesiology* 2008;**108**(1):71–7.
2. Kurz A, Sessler DI, Lenhardt R. Perioperative normothermia to reduce the incidence of surgical-wound infection and shorten hospitalization. Study of Wound Infection and Temperature Group. *N Engl J Med* 1996 May 9;**334**(19): 1209–15.
3. Frank SM, Fleisher LA, Breslow MJ, Higgins MS, Olson KF, Kelly S, et al. Perioperative maintenance of normothermia reduces the incidence of morbid cardiac events. A randomized clinical trial. *JAMA* 1997 April 9, 1997;**277**(14): 1127–34.
4. Scott EM, Leaper DJ, Clark M, Kelly PJ. Effects of warming therapy on pressure ulcers–a randomized trial. *AORN J* 2001 May;**73**(5):921–7, 9–33, 36–8.
5. Lenhardt R, Marker E, Goll V, Tschernich H, Kurz A, Sessler DI, et al. Mild intraoperative hypothermia prolongs postanesthetic recovery. *Anesthesiology* 1997 Dec;**87**(6): 1318–23.
6. NICE. Perioperative hypothermia (inadvertent): The management of inadvertent perioperative hypothermia in adults. London: National Institute for Health and Clinical Excellence; 2008 Contract No.: Document Number.
7. Camus Y, Delva E, Cohen S, Lienhart A. The effects of warming intravenous fluids on intraoperative hypothermia and postoperative shivering during prolonged abdominal surgery. *Acta Anaesthesiol Scand* 1996 Aug;**40**(7): 779–82.
8. Gelman S. Venous Function and Central Venous Pressure: A Physiologic Story. *Anesthesiology* 2008;**108**(4):735–48.
9. Bräuer A, English MJ, Steinmetz N, Lorenz N, Perl T, Weyland W, Quintel M. Efficacy of forced-air warming systems with full body blankets. *Can J Anaesth* 2007;**54**(1): 34–41.
10. Rein EB, Filtvedt M, Walloe L, Raeder JC. Hypothermia during laparotomy can be prevented by locally applied warm water and pulsating negative pressure. *Br J Anaesth* 2007 Mar;**98**(3): 331–6.
11. Turner M, Hodzovic I, Mapleson WW. Simulated clinical evaluation of four fluid warming devices. *Anaesthesia* 2006;**61**(6):571–5.

FURTHER READING

Sessler DI. Temperature monitoring and perioperative thermoregulation. *Anesthesiology* 2008;**109**(2):318–38.

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Chapter 31

Physics and technology of ultrasound

Crispian Oates

CHAPTER CONTENTS

What is ultrasound? 525

With the advent of low-cost portable scanning equipment, ultrasound has now become an important tool in the armoury of the anaesthetist. Such machines are becoming ever more sophisticated as functions once seen only on large, high-end equipment are migrating down to the entry level systems. In order to be able to use ultrasound equipment and interpret the images produced effectively and safely, it is important to understand some of the basic physical principles behind ultrasound as an imaging modality. This chapter has been included to cover that essential ground and to assist the practitioner to proficiently use ultrasound equipment and interpret what is seen.

WHAT IS ULTRASOUND?

Diagnostic ultrasound uses high-frequency sound waves sent into the body, to build up an image of the anatomical structures within from the detected echoes. A sound wave is what is known as a longitudinal compression wave (Fig. 31.1). It may be formed by some sort of piston, a loudspeaker for example, that pushes forward onto the molecules of air in front of it forming a region of higher density and pressure immediately in front of the piston. These molecules then push on to the molecules in front of them and so on, so the pulse of high pressure starts to move forward away from the piston. Meanwhile the piston moves backwards creating a region of low density and pressure in the molecules in front of it. By repeatedly moving forwards and backwards, a succession of high pressure followed by low-pressure regions move away from the piston forming a longitudinal pressure wave, a sound wave. If we were able to see just one molecule in front of the piston, we would see it oscillate forwards and backwards about an average position, like a child on a swing. Hence the molecules of the medium carrying the sound wave stay in approximately the same place, but the disturbance, i.e. the change in pressure, travels out, away from the piston.

The distance between two successive pressure peaks is called the wavelength (λ) of the sound wave, and the number of pressure peaks going past a fixed point in a second is the frequency (f). The speed with which a pressure peak moves forward is called the speed of sound (c). These three quantities are linked by a simple equation:

$$c = f\lambda$$

For any given medium the speed of sound is constant. This means that a high frequency implies a short wavelength and low frequency a long wavelength. Table 31.1 gives examples of the speed of sound travelling through different tissues, and also air and water. The speed of sound depends on how rigid a material is, that is: how strongly the molecules are bound together, and how massive the molecules are, i.e. how easy they are to move. In air the molecules are very poorly connected with one another and the speed of sound is low at 340 m s^{-1}. Bone is a very rigid material and the speed of sound is fast at 3000 m s^{-1}. Water has a speed of sound somewhere in between these figures: 1480 m s^{-1} at 20°C. The speed of sound for all the soft tissues in the body shows little variation, with values being not very different from that of water. For the purposes of image formation using ultrasound, the speed of sound in the different soft tissues can be approximated by an average value of 1540 m s^{-1} (+/− 10%) and this is the value of c that is used for ultrasound imaging.

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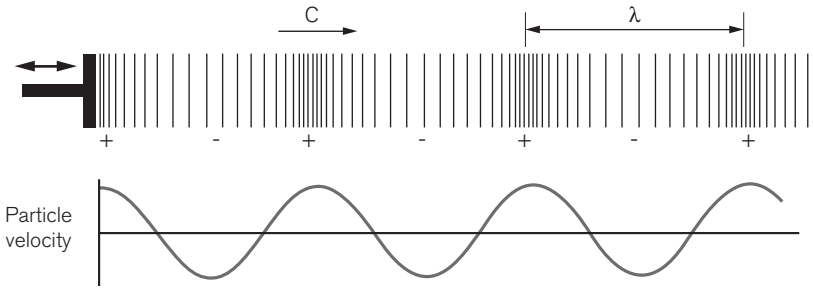


Figure 31.1 Soundwave showing particle velocity of molecules about their average position and pressure changes produced by piston source; c is speed of sound, λ is wavelength.

Table 31.1 The speed of sound in various media

SPEED OF SOUND m s^{-1}		
Blood	1570	Average 1540
Liver	1580	
Fat	1430	
Muscle	1575	
Bone	3000	
Air	340	
Water	1480	

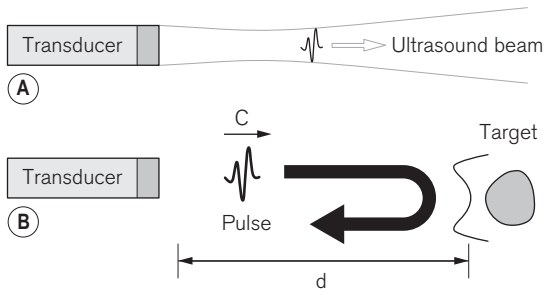


Figure 31.2 **A.** The ultrasound beam shape. **B.** The pulse-echo principle to determine target range; c is speed of sound, d distance to target.

Ultrasound is any sound with a frequency that is higher than the range of our hearing, which in a young person runs from 20 Hz to 20 kHz. For diagnostic imaging the frequencies used are far higher than anything we can hear and are in the range of 3–15 MHz. Putting these frequencies into the speed of sound equation, and using a speed of sound of 1540 m s^{-1}, we see that the wavelength of ultrasound in soft-tissue ranges from 0.5 to 0.1 mm. The reason for using these very high frequencies is to produce narrow beams of sound (see below) and to image with high resolution so that small structures within the body can be viewed.

Producing a narrow beam of sound is important. When we speak, the wavelength of sound in air is about 0.5 m and sound spreads out in all directions from our mouth, which is much smaller than that wavelength of sound. Using a very high frequency with a wavelength smaller than the transducer aperture produces a narrow well-defined beam of sound, like a narrow torch beam of light, enabling the ultrasound beam to be precisely pointed at the target of interest.

Pulse echo principle

Ultrasound imaging uses the pulse-echo principle in a similar manner to sonar on a submarine (Fig. 31.2). A short pulse of sound is sent into the body and echoes reflected off anatomical structures in the tissues return to the transducer. The received signals are used to build up an image of the anatomy. The pulse transmitted into the body is typically only two to three cycles long and lasts just less than 1 μs. If the transducer were to produce a continuous wave of sound there would be a beam of sound travelling into the body as described before (Fig. 31.2A). When a short pulse is used, the pulse travels down the same beam shape. In other words, the ultrasound beam can be thought of as the corridor down which the pulse travels and from which the echoes will arise. Any tissue lying outside the beam, in general will not see the pulse and not send any echoes back to the transducer. The pulse travels at 1540 m s^{-1} towards the target at some depth (d) in tissue. An echo is reflected from the target which travels back to the transducer at the same speed. The echo pulse has, therefore, travelled the distance twice, from the transducer to the target and back, and the time of arrival of the echo back at the transducer Δt is given by equation:

$$\Delta t = \frac{2d}{c}$$

For a speed of sound of 1540 ms^{-1}, the go and return time of the pulse is 13 μs per cm. Because the machine can assume a constant value of sound of 1540 m s^{-1}, the scanner can use the timing of echo arrival to position

echoes in the image and mark the depth axis of the display in centimetres depth.

The ultrasound transducer

The transducer plays the role of the piston in producing the sound wave (Fig. 31.3A). It converts electrical energy into sound energy and vice versa. Ultrasound transducers use a piezoelectric material to produce a sound wave. This is a material having the property that when an electrical voltage is placed across it, it changes shape and contracts in the direction of the applied voltage. If the polarity of the voltage is reversed, the piezoelectric material expands. The process is reciprocal. If the material is squeezed, a voltage forms across it, and if the material is stretched, the voltage produced has the opposite polarity. A naturally occurring example of a piezoelectric material is quartz. This is what forms the resonator in a quartz clock. Ultrasound transducers usually use a ceramic material called lead zirconate titanate (PZT). A piece of this material is used whose thickness is such that it has a natural resonance at the frequency of ultrasound we wish to use, say 10 MHz. The PZT element has electrodes on its faces. When a high-voltage spike is applied to the electrodes the PZT is made to rapidly contract. It then vibrates at its resonant frequency, like an empty wine glass struck with a fingernail. In order to make a very short pulse the back of the PZT element has a layer of sound absorbing material (*backing layer*) stuck to it to reduce the ringing (Fig. 31.3B). This shortens the pulse in a similar way to striking the wine glass, but with the fingers of the other hand pinching the rim of the glass.

The front of the transducer has a '*matching layer*' on it that fulfils a similar function to the coloured coatings seen on camera lenses. It reduces reflection at the surface and enables the ultrasound pulse and echoes to get into and out of the body more efficiently. The front of the transducer then has a polymer *lens* stuck to it to focus the beam and improve resolution.

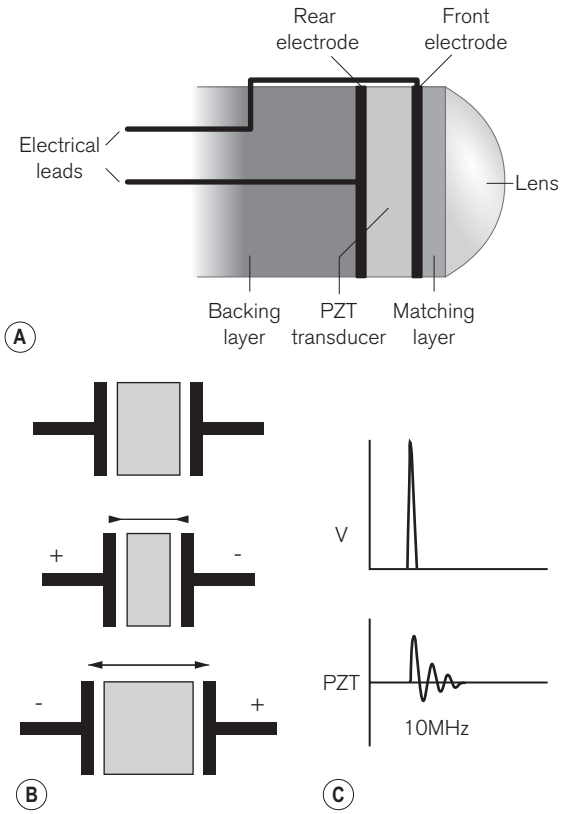


Figure 31.3 **A.** Diagram of a typical transducer construction. **B.** The characteristic of a piezoelectric material showing relationship between shape change and associated voltage across the element. **C.** Damped resonant pulse from PZT transducer element following applied voltage spike.

Image formation

The way a modern ultrasound scanner forms an image may be considered in three stages. Firstly, if an ultrasound beam is directed into the body towards a series of reflecting targets the output from the received echoes can be displayed as a series of peaks whose time of arrival depends on the depth equation and whose amplitude depends on the reflecting strength of the targets. A strong reflector gives a high amplitude signal and a weak reflector gives a small signal. This type of display is known as an *A-mode* or *amplitude mode* display (Fig. 31.4).

The second step is to represent the amplitude of the targets, not as an oscilloscope type display, but as shaded spots along a single line. The spots are shaded according to a grey scale with white representing the strongest reflection, black the weakest or no signal, and other shades of grey for signals whose strength lies in between. This display is the same as one scan line from the transducer into tissue in a full ultrasound image.

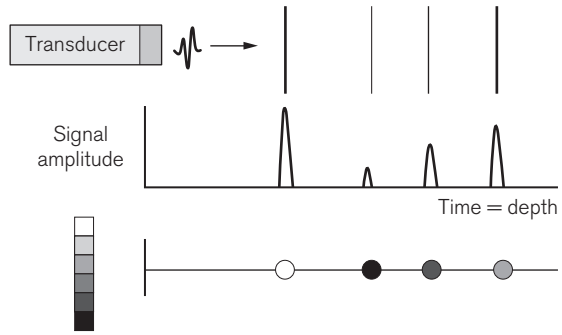


Figure 31.4 The formation of one image line from one transmitted ultrasound pulse-echo sequence.

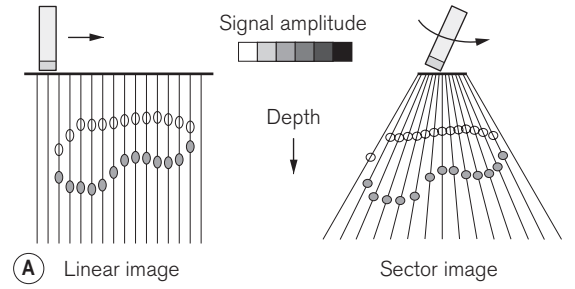




Figure 31.5 **A.** Principle of formation of linear and sector image formats. **B.** Examples of linear and curvilinear array transducers, producing linear and sector format displays respectively.

The final step is to sweep the beam through the tissue and to repeatedly interrogate the tissue with a series of pulses, displaying the resulting grey scale encoded lines side by side so as to form an image of extended targets in tissue. This type of display is known as a *B-mode*, for *brightness mode* or *grey scale image* (Fig. 31.5A).

If the transducer is moved across the skin in a line, a rectangular image is produced known as a *linear scan*. If the ultrasound transducer is rotated at one point on the skin so as to sweep the ultrasound beam through the tissue in an arc, the image has the shape of the sector of a circle and is known as a *sector scan*. Each scan format has its uses, for example, the sector scan has a small footprint on the skin that spreads out inside the body, so it is used for echocardiography where the beam has to get in between the ribs, but the whole heart can be seen below.

Modern transducer probes consist of an array of 128–256 piezoelectric elements mounted in a flat line to form a linear array, or on a curve to produce a curvilinear array, producing linear and sector images respectively (Fig. 31.5). Groups of elements can be switched electronically between transmit–receive pulse sequences to move the active area along the array. In this way a series of scan lines is

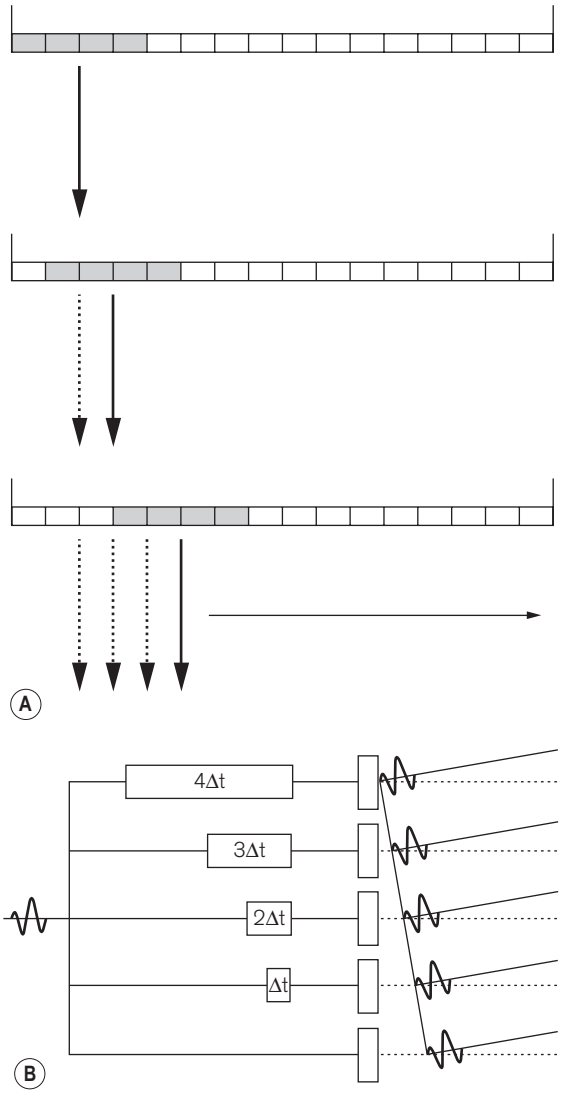


Figure 31.6 **A.** Principle of electronically scanning an image by stepping an active set of elements across the array for successive pulses. **B.** Electronic beam steering achieved by introducing time delays into the path between the summing point and the elements.

produced, scanning through the target to form an image (Fig. 31.6A). The whole image is then scanned and refreshed many times each second to give a continuous real-time image that can show any movement taking place, for example a needle moving towards its intended target.

The beam may be steered through different angles by inserting time delays in front of individual elements (Fig. 31.6B). Because the pulse then arrives at adjacent elements at successively later times, the wavefronts of the soundwave propagating away from the elements add up

along a line at an angle to the face of the probe. In other words, the transmitted pulse is steered away from a line perpendicular to the probe face. When echoes return to the transducer, the received echoes are again delayed and then added together. Those echoes coming from the steering angle will give a strong signal, as all their wave crests will add together. But for those echoes arriving from other angles, the peaks and troughs will tend to cancel each other out and a signal will not be detected. By switching the time delays electronically, each successive pulse can be steered at a different angle to produce a sector-shaped image. This is the basis for the function of *phased array probes*, commonly used for cardiac studies and for some intra cavitary applications.

The ultrasound journey

As the ultrasound pulse travels through the body it interacts with the tissue in various ways (Fig. 31.7). At large smooth interfaces, such as are found between one organ and another, muscle–liver for example, some of the energy in the pulse is reflected back towards the transducer as an echo (Fig. 31.7A). This type of reflection from a large smooth interface is known as *specular reflection* from the Latin *speculum* meaning a mirror. The strength of the echo will depend on the speed of sound (c) and density (ρ) of the tissues on either side of the interface through a property known as characteristic acoustic impedance (z_o). I_i is the incident intensity and I_r is the reflected intensity of ultrasound (Table 31.2).

$$Z_0 = \rho c$$

$$\frac{I_r}{I_i} = \left[\frac{Z_2 - Z_1}{Z_2 + Z_1}\right]^2$$

In practice this means that in going from one soft tissue to another, e.g. muscle–fat, where the density and speed of sound are very similar, only ~1% of the sound energy in the pulse is reflected back to the transducer. The rest of the pulse energy is transmitted on to deeper tissues. In going from soft tissue to a gas there is a very big difference in acoustic impedance between the two. Gas has a very low density and low speed of sound compared with soft tissue. In this case 99.9% of the sound energy is reflected back at the first tissue–gas interface and no ultrasound is transmitted to deeper tissues. The image will show a very bright reflection with no information beyond it. This is also the reason why it is necessary to put gel on the skin of the patient, to exclude all air between the transducer

Table 31.2 Reflectivity

Soft tissue–soft tissue	~1%
Soft tissue–air	99.9%
Soft tissue–bone	~50%

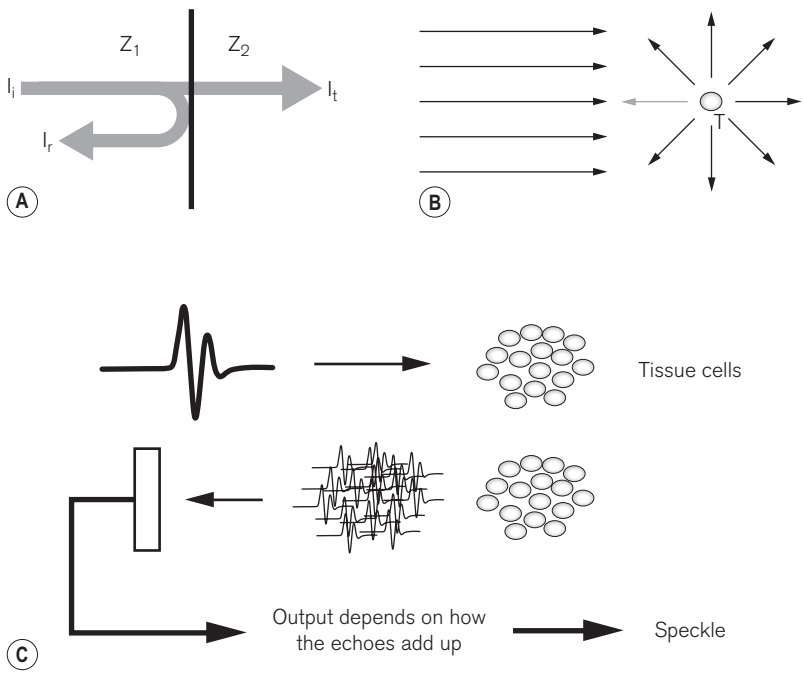


Figure 31.7 **A.** Specular reflection at a large interface. **B.** Rayleigh scattering at a target smaller than a wavelength of ultrasound. **C.** Principle of speckle formation from superposition of scattered pulses.

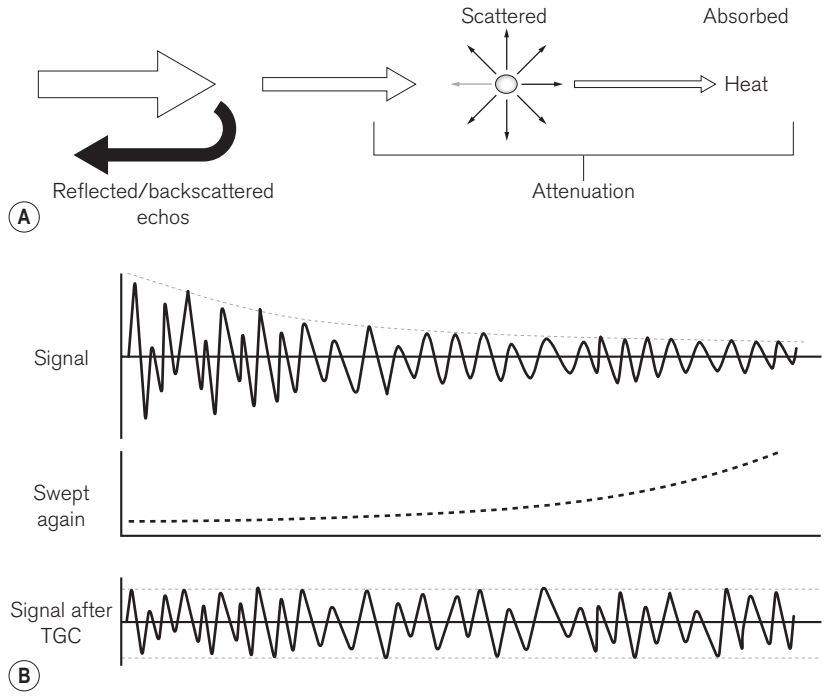


Figure 31.9 **A.** Attenuation of the ultrasound pulse as it travels through tissue. **B.** Principle of time gain compensation to negate the effects of attenuation.

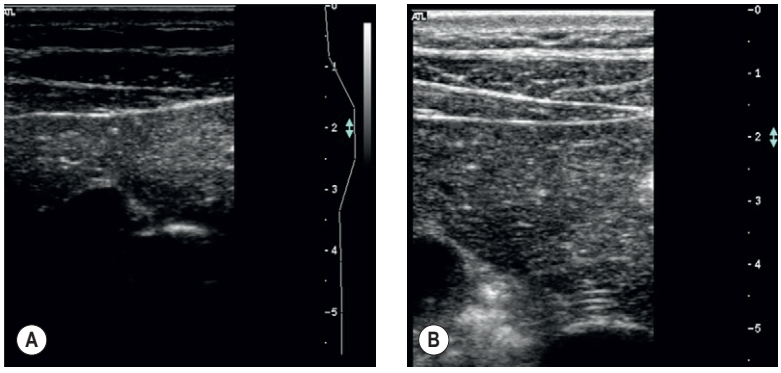


Figure 31.10 **A.** Time gain compensation incorrectly adjusted and **B.** correctly adjusted. Note the overall gain is set to just show peak white on the strongest echoes.

In order to overcome this problem the ultrasound scanner applies a gradually increasing gain to the received signal from each transmitted pulse. By this means the decrease in signal strength from echoes arriving later in time, from greater depths, is compensated for by amplifying the signal more. This technique is known as *time gain compensation* (TGC) or *depth gain compensation*. It does not increase the penetration of the ultrasound probe for when attenuation has extinguished the pulse, there are no more echoes returning. However, it does enable the image over the useful depth to be uniformly displayed. Beyond the penetration depth only amplified noise will be displayed on the image. TGC is a control that must be correctly adjusted by the user to ensure information in the image is not missed or misinterpreted. Most scanners apply a basic TGC which the user can further adjust. On larger machines the TGC control is usually in the form of sliders set to control the image brightness at 1 cm steps in depth. On portable scanners the control is often simplified to a near field and a far field gain. There is also an overall B-mode gain control affecting the whole image. This should be adjusted so that the brightest parts of the image just reach peak white (Fig. 31.10).

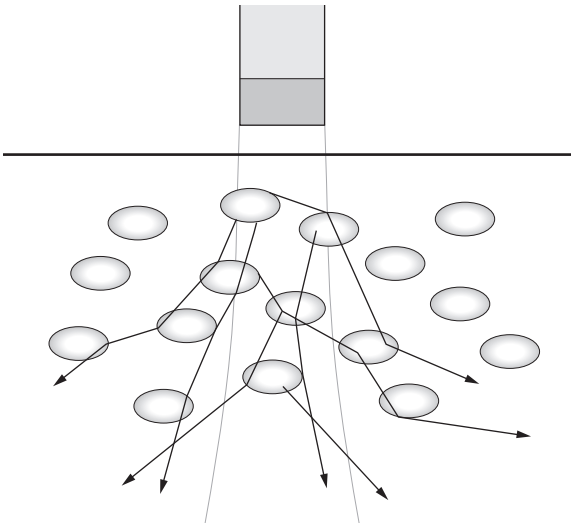


Figure 31.11 Diagram illustrating how beam distortion occurs with certain types of body fat to produce poor visualization.

Poor visualization

In some patients the image will appear somewhat murky and poorly defined and may be described as *poor visualization*. This is a result of some types of body fat distorting what should be a well-defined ultrasound beam (Fig. 31.11). Fat consisting of slightly more dense fat globules in a matrix of less dense fat has the same effect on an ultrasound beam as the distorting glass used in bathroom windows has on light. It depends on the particular patient and when seen there is little the operator can do. Pressing the transducer harder on the skin or trying an alternative line of approach to the target may help, but in the end it is one of the limitations of ultrasound as an imaging modality.

Resolution

Resolution is a measure of how fine a detail can be seen in an image and answers the question: '*How close can two targets be and still be seen as two separate targets rather than one large blob on the image*?' It needs to be considered in two planes, along the axis of the ultrasound beam and across the width of the ultrasound beam.

Axial resolution (Fig. 31.12A) depends on the length of the pulse travelling out along the ultrasound beam. If two targets along the beam axis are just within the pulse at the same time, the echo sent back to the transducer from one target will just overlap the other. In order to see the two targets separately they must be at least half a pulse length apart. The pulse length is $n\lambda$, where n is the number of cycles in the pulse and λ the wavelength of ultrasound. A

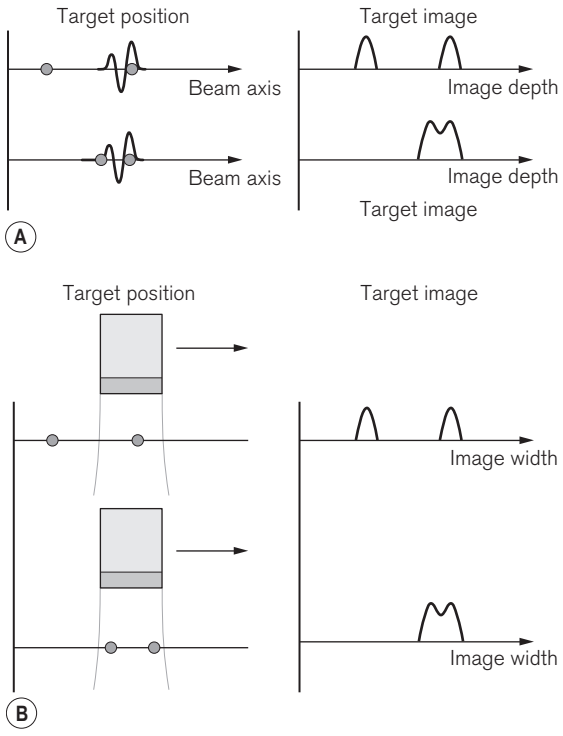


Figure 31.12 **A.** Axial resolution showing position of targets at different depths to be resolved. **B.** Lateral resolution showing position of targets relative to beam width to be resolved.

typical transmitted pulse is three cycles long. This gives a pulse length of 0.45 mm at 10 MHz and, therefore, an axial resolution of 0.23 mm. In order to improve the resolution it is necessary to use a shorter pulse or a higher frequency giving a shorter wavelength.

Lateral resolution (Fig. 31.12B) is the resolution in the direction across the beam width, i.e. orthogonal (at right angles) to the axial resolution. As the ultrasound beam is scanned across a target it will just be separated from an adjacent target if the adjacent target is just leaving the beam as the first target is entering the beam. They will be seen as two separate targets on the image if they are more than half a beam width apart. Beam width also depends on frequency with a narrow beam being produced by a higher frequency for a given aperture or active area on the transducer array. For a 10 MHz probe lateral resolution will be slightly greater than the axial resolution at 0.4 mm. From this discussion it may be seen that both the axial and lateral resolutions are improved at higher frequencies, hence the desire to use high-frequency probes.

The lateral resolution can be further improved if a lens is put in the ultrasound beam to focus it in a similar way to putting a lens in front of a torch to focus its light (Fig. 31.13A):

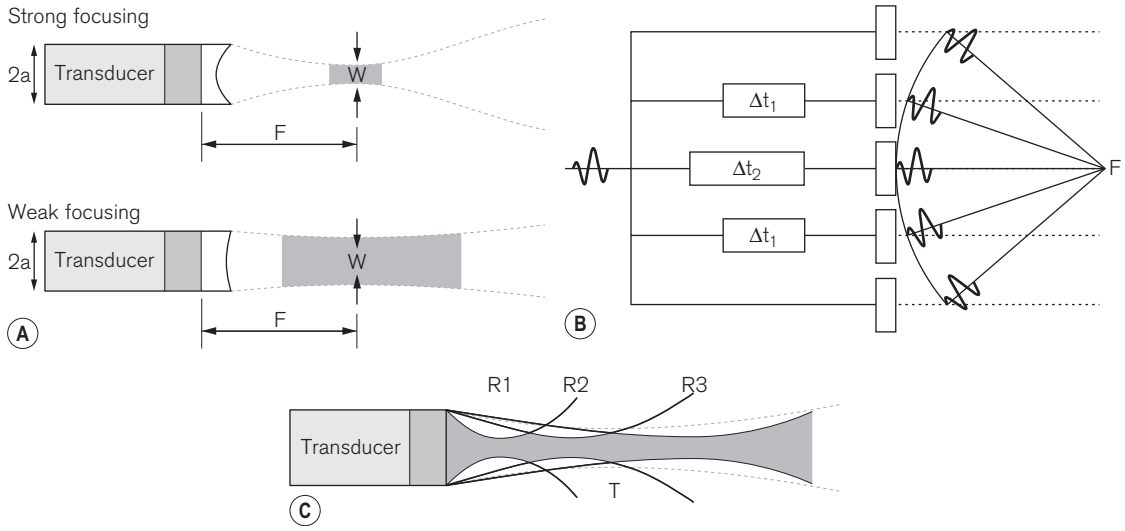


Figure 31.13 **A.** The effect of focusing an ultrasound beam. The shaded area is the focal zone, where resolution is improved by focusing. Strong focusing gives a narrower beam width, but a shorter focal zone than weak focusing. Note also the increased divergence of the beam in the far zone for strong focusing compared to weak focusing (a = half width of transducer aperture, see text). **B.** Electronic focusing using time delays to produce a wave converging onto F the focal point. **C.** Ultrasound beam shape with one focal zone in transmit T and multiple focal zones on receive. Time delays to focus at R1, R2 and R3 are switched in when echoes are arriving from those depths.

$$w = \frac{F\lambda}{a}$$

where w is the beam width at the focus, F is the distance of the focus from the transducer, λ is the wavelength and a is the half-width of the transducer aperture.

Focusing produces a narrower beam than if no focusing were used over a range along the axis known as the *focal zone*. However, the beam width outside the focal zone is wider than if no focusing were to be applied, as the beam spreads out more. The stronger the focusing, the narrower the focal zone, but, in addition, the worse the beam spread outside the focal zone.

Electronic focusing

An ultrasound beam will be focused if all of the wavefronts in the sound wave converge on a single point in space: the *focal point*. This may be achieved by a lens, which refracts or bends the wavefronts into such a converging wave. When a transducer array is used, as is the case in medical imaging, focusing can be achieved electronically by the use of time delays, in a similar manner to electronic beam steering (Fig. 31.13B). Time delays are chosen to control when the transmitted pulse reaches each element so that the propagating wavefront forms a converging line along a circle centred on the desired focal point in front of the transducer. As with the previously described beam steering, the peaks and troughs of the received echoes will add up strongly for echoes arising from the beam directed toward the focal point, but tend to cancel for other directions. By changing the time delays, it is possible to focus the beam on different depths in the tissue. The time delays can be rapidly switched from one set to another so that multiple focal zones can be used in receive, improving the resolution at all depths (Fig. 31.13C). That is, for each transmitted pulse, as echoes arrive from successively greater depths, the next set of time delays is switched in order to focus at that depth. For the transmitted pulse itself, only one focal zone can be used, as that focused pulse travels out to all depths of tissue before it is fully attenuated. Once the pulse has left the transducer, nothing further can be done to change its shape or direction.

On the larger scanners it is possible for the user to choose at what depth of tissue to place the *transmit* focal zone. Its position is usually shown by a marker down the side of the image. Because the lateral resolution is poorer outside the focal zone, it is important to ensure that the focal zone is set to the depth of the target of interest when scanning. On small portable scanners a weaker fixed focus (that the user has no control of) is used, giving some improvement in lateral resolution over the central area of the image.

Slice thickness

When viewing an ultrasound image we see a slice through the body (Fig. 31.14). The slice in the body is called the *scan plane* or *image plane*. It is important to note that the scan plane slice is not infinitesimally thin. The ultrasound

beam has a significant width in the out-of-plane or elevation direction. This may be 2–3 mm for a 10 Mhz transducer. This is known as *slice thickness* and tissue that is outside the image plane, but still within the elevation plane beam width, will also send echoes back to the transducer. This contributes to some degradation in the sharpness of image. It is minimized by the manufacturer putting a weak cylindrical lens across the front of the transducer probe to give some mild focusing in the elevation plane direction.

Artefacts

When performing ultrasound imaging it is very important to learn to recognize the artefacts that may occur in an image. An artefact is any feature in an image that is not an exact one-to-one representation of the target being imaged. Every imaging modality has artefacts and ultrasound imaging has them resulting from the fact that pulses of sound in a beam of finite width are bouncing around inside the body, through different tissue types whose speed of sound is not exactly the same. Some artefacts actually prove to be useful in helping to identify particular types of targets or anatomical structures. In order to make a correct interpretation of the image it is necessary to make sure that: the images are correctly oriented, the probe is the right way round in your hand, you know and recognize the anatomy of the target you are viewing, and you recognize and understand any artefacts there may be in the image.

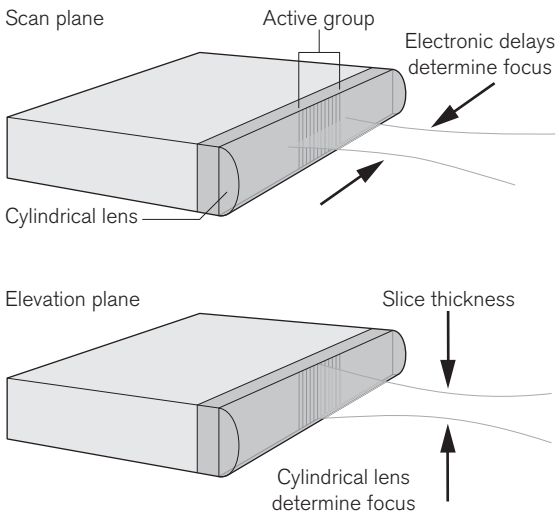


Figure 31.14 Diagram to show the scan plane and the elevation plane beam shapes of a transducer. The elevation plane beam width is the slice thickness.

Shadowing (Fig. 31.15)

Where an ultrasound beam reflects from gas or a dense solid target like bone there will be an acoustic shadow behind the obstacle similar to shadow production in a beam of light. Firstly, this gives some information on the nature of the target; for example, calcified plaque will have a bright surface with shadowing behind. Secondly, there is a loss of information within the shadow and the probe may have to be repositioned to see what is there. Where an artery is viewed in cross-section, the walls parallel to the axis of the beam will not be imaged. The sound is deflected by the shallow angle of the fibrous adventitia around the tangent to the beam axis and there will be shadowing behind it.

Post cystic enhancement (Fig. 31.15)

Clear fluids are poor scatterers of ultrasound and there is very little attenuation of the ultrasound as it travels through them. They therefore appear black on the ultrasound image. However, the TGC is operating on the basis that the ultrasound is being attenuated as it travels through tissue. The consequence is that behind the fluid-filled structure, the image is brighter than it should be because the

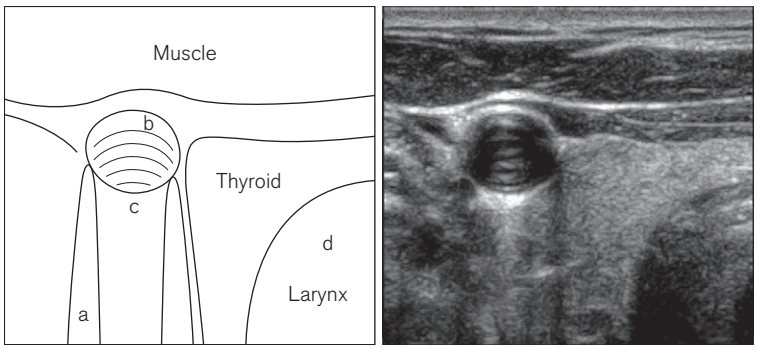


Figure 31.15 Image and schematic of a cross-section through the common carotid artery and thyroid showing: **a,** shadowing by arterial walls, **b,** reverberation within the artery from fascia and the arterial wall; **c,** post-cystic enhancement distal to the artery. Note, everything seen in the larynx **d,** is artefact as it is air-filled.

gain has been increased, but the attenuation has not taken place. This effect is known as post cystic enhancement and, together with the dark image of the fluid, is confirmation that the structure seen in the image is fluid filled, such as a cyst, a blood vessel or other fluid collection.

Reverberation (Fig. 31.15)

Where parallel reflectors occur in the image plane perpendicular to the beam axis, there is the possibility of pulses reflecting two or more times between them. This gives rise to reverberation artefacts. The pulses that have travelled twice between reflecting layers before going back to the transducer will arrive later and so will be misplaced in the image, appearing as duplicates of the real structure at greater depths in the image. There may be several of these depending on how many times the pulse bounces between the parallel reflectors. The reflection may be between an interface and the front surface of the probe, in which case the reverberation artefact will be at twice the distance of the interface on the image.

Multiple reflections may also occur within a single target. This will be the case if a hard solid target starts to ring like a bell when it is hit by an ultrasound pulse. The effect is seen on the image as a bright streak or comma behind the target along the beam axis. This is a useful method to detect foreign bodies in tissue and to spot where a needle is, from the reverberation comma seen (Fig. 31.16).

Mirror artefact (Fig. 31.17)

Specular reflection of the ultrasound beam by an extended interface can give rise to a reflection or mirror artefact. This results in a duplicate of one part of the target being misplaced over another area in the image. An example is seen when imaging the liver with the probe angled cephalad toward the diaphragm. A mirror image of the liver is seen in the air-filled space above the diaphragm as the ultrasound beam changes direction, reflecting back into the liver, whilst the scanner continues to draw a straight image line down the screen. The echoes from the reflected beam in the liver are, therefore, misplaced above the diaphragm in the lung space. Another situation where reflection artefact is seen is when looking down into the supraclavicular space where the blood vessels may be seen to be duplicated due to reflection in the top surface of the lung.

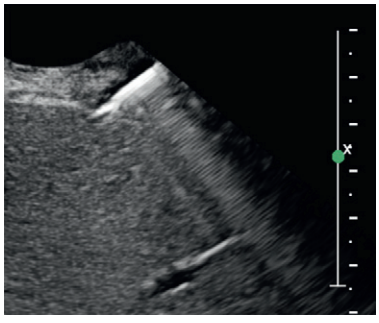


Figure 31.16 Reverberation caused by a biopsy needle.

Slice thickness artefact (Fig 31.18)

As described above, the scan plane from a real transducer is not infinitesimally thin, but has a beam width thickness. Apart from giving some general reduction in sharpness of the image, this can give rise to a specific artefact seen when viewing curved surfaces such as a longitudinal view of an artery. As the arterial wall curves away in the elevation plane direction, it is still within the beam of the transducer and so still sends back echoes. This appears on the image as a thickening or extension of the vessel wall into the blood-filled vessel lumen, which would normally appear black on the image.

Refraction (Fig. 31.19)

Throughout the image forming process in soft tissue, the assumption is made that the speed of sound is constant at 1540 m s^{-1}. The fact that the speed of sound in real tissue varies by a small amount between the different tissue types gives rise to refraction artefacts. When the ultrasound beam crosses a surface obliquely, the difference in speed of sound either side of the interface causes the beam to be bent, leading to distortion in the image. This is the same effect as seen optically when a pencil is put into a glass of water and viewed from the side. The pencil appears to bend at the air–water interface due to the difference in the speed of light in air and water. Any distortion caused in an ultrasound image is not usually noticed,

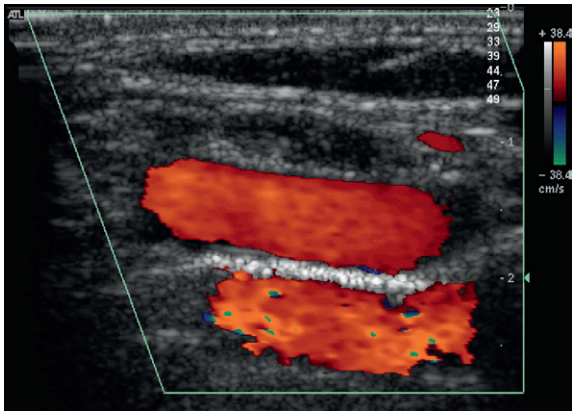


Figure 31.17 An example of mirror artefact. The subclavian artery is duplicated due to reflection from the upper surface of the lung in the supra-clavicular view.

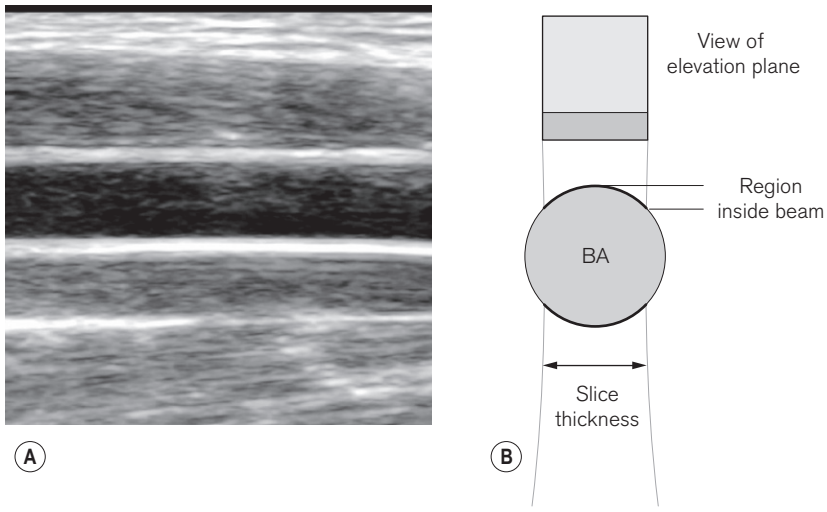


Figure 31.18 **A.** Brachial artery (BA) aligned correctly to show bright walls with the intimal layer just seen. The scan plane then passes through the centre-line of the vessel. The echoes seen below the proximal wall are due to slice thickness artefact. They are an image of the arterial wall picked up in the elevation plane as shown in **B**.

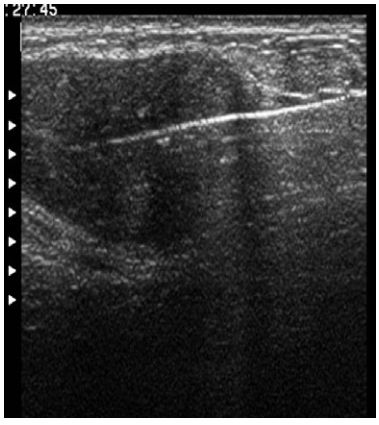


Figure 31.19 Refraction artefact seen in what appears to be a bent needle aspirating a cyst.

but when a straight needle is put into tissue and viewed with ultrasound, the needle may appear to be bent as it passes through different tissue layers.

Viewing anatomical structures

In order to produce a bright sharp surface of an anatomical structure in an image, it is necessary to angle the ultrasound probe so that the ultrasound beams insonate the surface of that structure at 90° (Fig. 31.18). As the surface curves away from the perpendicular it becomes less distinct and slice thickness, reflection and shadowing artefacts become more apparent. Hence if a blood vessel is viewed in longitudinal section, when the image of the walls is sharp and bright, the ultrasound beam is being directed through the centre-line of the vessel. Knowing this can aid ultrasound-directed insertion of needles and cannulas. The easiest way to obtain this longitudinal view is to image the vessel in cross-section with the vessel in the centre of the field of view. Then rotate the probe into the longitudinal view keeping the vessel at the centre of the image as you do so.

Some tissues have a particular orientation of their structural components causing the ultrasound appearance to vary, depending on the orientation of the image. Their ultrasound appearance is then said to be *anisotropic*. For example, anisotropy is seen in muscle which has a very strongly striated appearance when viewed along the muscle fibres, but a more uniform appearance when viewed across the fibres. Tendons and nerves also show anisotropy, this being more pronounced in tendons. The effect can be to make the target difficult to see if the probe orientation is not ideal (Fig. 31.20).

Because of the large difference in acoustic impedance between gas and soft tissue, the introduction of any bubbles, even microscopic bubbles, from any fluid injected into the tissue, will show up on the image as a bright speckled reflection and will obscure the image from greater depths.

Compound mode

Some scanners have a *CT* or compound scanning mode that may be turned on or off by the user (Fig. 31.21). This mode sends ultrasound beams into the body at several angles to the face of the transducer from across the whole

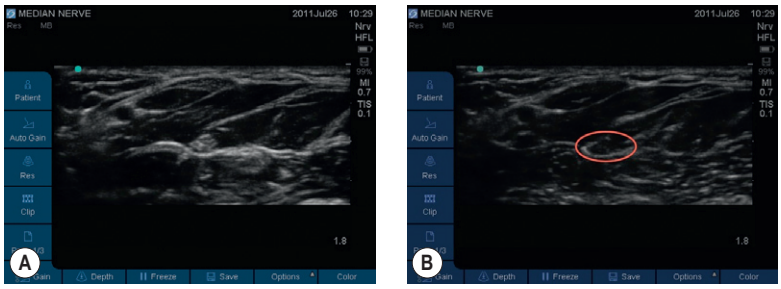


Figure 31.20 Two images of the median nerve in the forearm showing the effect of its anisotropy. With slight alteration of the probe angle the nerve (marked in B) becomes less visible.

width of the probe. The effect of this is to insonate each target from a range of angles so reducing the effect of shadowing by sending beams in behind the obstacle, and showing the curved surface of rounded structures more uniformly as the surface is seen by a beam insonating it perpendicularly at more points. Use of compound mode may reduce frame rate and give some blurring of edges due to misregistration of images from multiple directions.

Harmonic imaging

Some scanners have a harmonic imaging mode that the user can switch on or off (Fig. 31.22). This mode reduces the effects of some artefacts, lessening what is often called *clutter* in the image, and gives some improvement in resolution. It uses the fact that as an ultrasound pulse travels through tissue it changes shape in a way analogous to

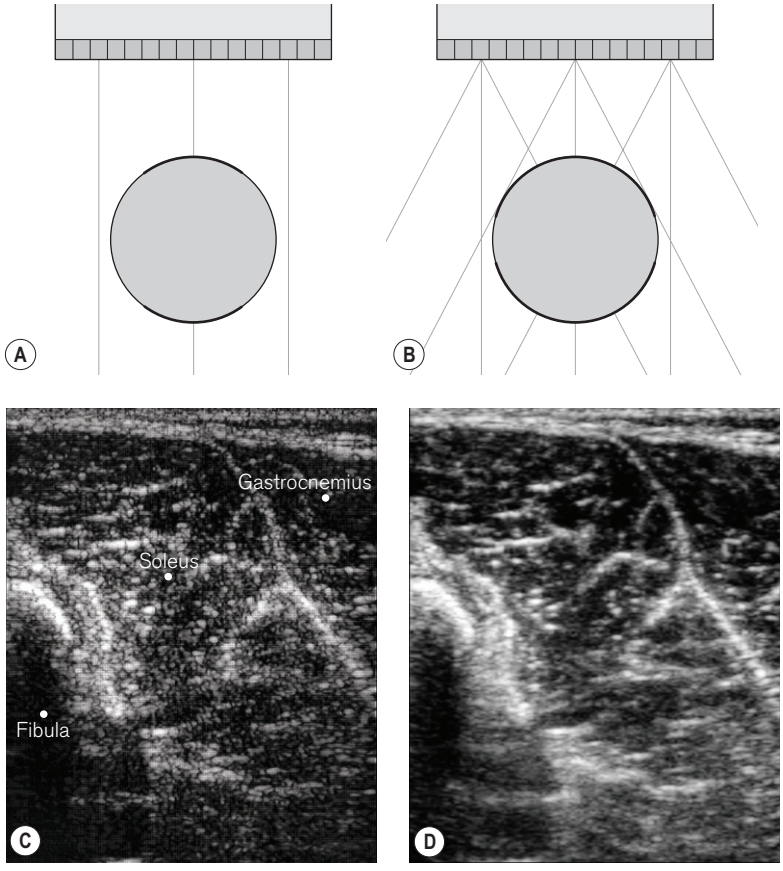


Figure 31.21 Increased visualisation with compound mode (**B** and **D**) showing the effect on the lateral calf muscles.

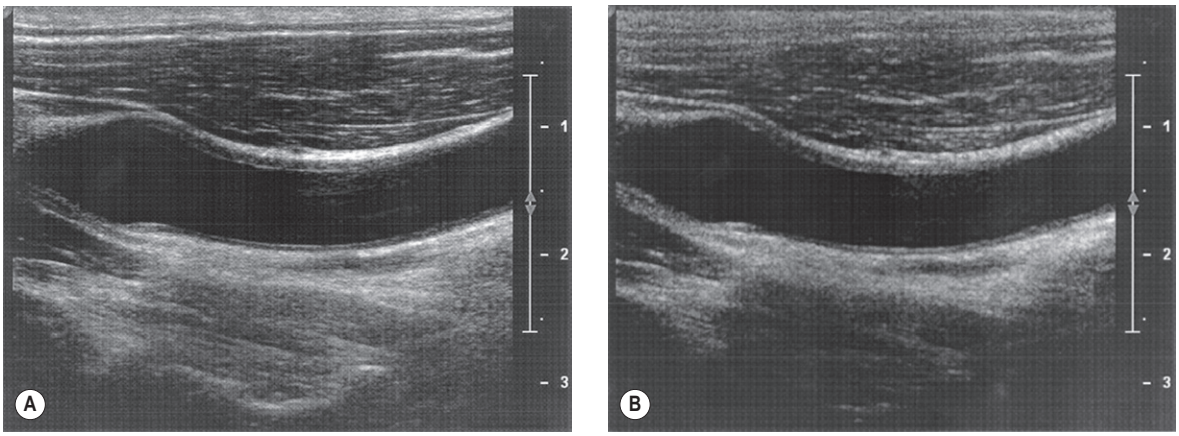


Figure 31.22 Illustration of the effect of harmonic imaging showing a longitudinal view of the common carotid artery. **A.** Harmonic imaging off. **B.** Harmonic imaging on.

a wave on the seashore changing its shape as it begins to break. The peaks become more pointed compared with the troughs. What is happening is that some of the energy of the pulse is going into a higher frequency at double the fundamental frequency of the pulse wave. So, for example, if the pulse has a fundamental frequency of 4 MHz when it is transmitted, some of the energy moves to a frequency of 8 MHz as the pulse passes through tissue. If the transducer is a wideband transducer, it can detect both 4 MHz and 8 MHz echoes. In harmonic imaging mode a high-pass filter is used so that the higher frequency 8 MHz echoes are detected and the 4 MHz component is removed. This improves the resolution as the imaging frequency is effectively increased. When using harmonic imaging, the penetration is often poorer and image contrast is reduced.

Colour Doppler ultrasound

The Doppler effect is very familiar to anyone who has stood at the side of the road and heard a fast police car pass by with its siren going. The Doppler effect is the change in the pitch of the siren heard as the sound waves are squashed together to give a higher frequency as the source comes towards the 'observer' and stretched out to give a lower frequency as it moves away. The same effect is used to image moving targets within the body. In particular it is very useful for detecting blood vessels and to aid identification of arteries and veins according to the direction and pulsatility of flow within them. Colour Doppler ultrasound (CDU) uses the difference between the transmitted frequency and the received frequency of the echoes to colour the B-mode image with an overlay, or colour map, showing where there is movement (Fig. 31.23). The difference between received and transmitted frequencies is known as the *Doppler frequency*. A colour

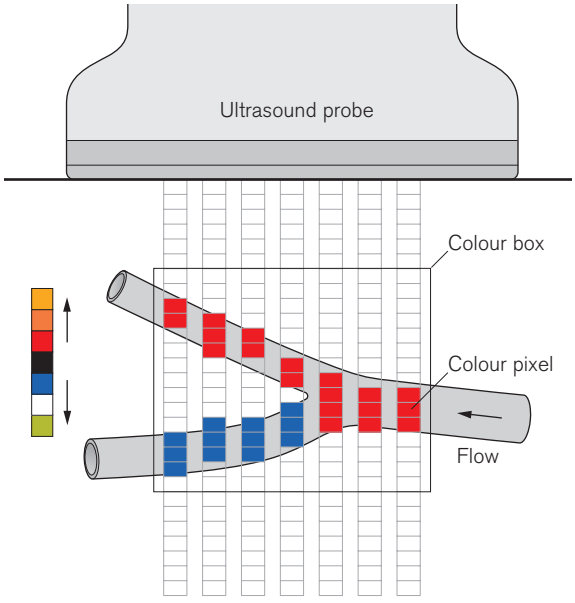


Figure 31.23 Diagram showing the overlay mapping of velocity used in colour Doppler ultrasound.

scale with red-yellow for increasing velocity in one direction and blue-green in the other is used. It requires there to be an angle between the blood flow and the ultrasound beam as no Doppler signal will be detected if the vessel is at 90° to the ultrasound beam. The colour map is usually shown within a colour box on the image whose size and angle may be changed, causing angled ultrasound beams to be sent out to avoid the 90° no-signal problem on a

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Appendix

SI units and conversion tables

Chetan Patel

SI UNITS

In 1960, the Conférence Générale des Poids et Mesures (11th CGPM), which is the international authority on the metric system, replaced all previous systems with the modern metric system officially named the Système International d'Unités, abbreviated to SI. Over the years, this metric system has been adopted throughout the world and is continually being revised to parallel developments in science and technology.

The SI is founded on seven base units (Table App. 1). Other derived units are defined in terms of the seven base units and, for ease of understanding and convenience, may have special or compound names and their own symbols (Tables App. 2, App. 3). There are also 20 SI prefixes used to form decimal multiples and submultiples (Table App. 4).

Table App. 1 The seven base units of the SI

PHYSICAL QUANTITY	BASE UNIT	SYMBOL
Amount of substance	mole	mol
Electric current	ampere	A
Length	metre	m
Luminous intensity	candela	cd
Mass	kilogram	kg
Thermodynamic temperature	kelvin	K
Time	second	s

Table App. 2 Examples of SI-derived units with special names

PHYSICAL QUANTITY	UNIT NAME	SYMBOL	EXPRESSED IN BASE UNITS
Plane angle	radian	rad	$m\ m^{-1} = 1$
Solid angle	steradian	sr	$m^2\ m^{-2} = 1$
Frequency	hertz	Hz	s^{-1}
Force	Newton	N	$m\ kg\ s^{-2}$
Work/energy	joule	J	$m^2\ kg\ s^{-2}$ (N m)
Pressure/stress	pascal	Pa	$m^{-1}\ kg\ s^{-2}$ (N m^{-2})
Power	watt	W	$m^2\ kg\ s^{-3}$ (J s^{-1})
Electric charge	coulomb	C	s A
Electric potential difference	volt	V	$m^2\ kg\ s^{-3}\ A^{-1}$ (W A^{-1})
Electric capacitance	farad	F	$m^{-2}\ kg^{-1}\ s^4\ A^2$ (C V^{-1})
Electric resistance	ohm	Ω	$m^2\ kg\ s^{-3}\ A^{-2}$ [V/A]
Electric conductance	siemens	S	$m^{-2}\ kg^{-1}\ s^3\ A^2$ [A/V]
Magnetic flux	weber	Wb	$m^2\ kg\ s^{-2}\ A^{-1}$
Magnetic induction	tesla	T	$kg\ s^{-2}\ A^{-1}$ (Wb m^{-2})
Inductance	henry	H	$m^2\ kg\ s^{-2}\ A^{-2}$ (Wb A^{-1})
Luminous flux	lumen	Im	cd sr
Illuminance	lux	Ix	cd sr m^{-2}

Table App. 3 SI units with compound names

PHYSICAL QUANTITY	NAME OF UNIT	SYMBOL
Area	square metre	m^2
Volume	cubic metre	m^3
Speed/velocity	metre per second	$m\ s^{-1}$
Acceleration	metre per second squared	$m\ s^{-2}$
Density	kilogram per cubic metre	$kg\ m^{-3}$
Electric field strength	volt per metre	$V\ m^{-1}$
Magnetic field strength	ampere per metre	$A\ m^{-1}$
Current density	ampere per metre squared	$A\ m^{-2}$
Specific heat capacity	joule per kilogram per kelvin	$J\ kg^{-1}\ K^{-1}$

Table App. 4 The 20 SI prefixes

PREFIX	SYMBOL	ORDINARY NOTATION	10^x
yotta-	Y	1 000 000 000 000 000 000 000 000	10^{24}
zetta-	Z	1 000 000 000 000 000 000 000	10^{21}
exa-	E	1 000 000 000 000 000 000	10^{18}
peta-	P	1 000 000 000 000 000	10^{15}
tera-	T	1 000 000 000 000	10^{12}
giga-	G	1 000 000 000	10^9
mega-	M	1 000 000	10^6
kilo-	k	1000	10^3
hecto-	h	100	10^2
deca-	da	10	10^1
		1	10^0
deci-	d	0.1	10^{-1}
centi-	c	0.01	10^{-2}
milli-	m	0.001	10^{-3}
micro-	μ	0.000 001	10^{-6}
nano-	n	0.000 000 001	10^{-9}
pico-	P	0.000 000 000 001	10^{-12}
femto-	f	0.000 000 000 000 001	10^{-15}
atto-	a	0.000 000 000 000 000 001	10^{-18}
zepto-	z	0.000 000 000 000 000 000 001	10^{-21}
yocto-	y	0.000 000 000 000 000 000 000 001	10^{-24}

CONVERSION TABLES

French/Gauge

By convention single lumen catheters, like needles, are identified by Gauge (G) as a contraction for Imperial Standard Wire Gauge (SWG) (Table App. 5). Multi-lumen catheters and other larger diameter devices (particularly where there is a non-circular cross-section, e.g. double lumen tracheal tubes) are labelled by French gauge (F or FG). French gauge is also referred to as Charriere gauge (CH), after the 19th century Parisian surgical instrument maker, and corresponds to the external circumference in millimetres, which approximately equates to three times the maximal external 'diameter'.

Each lumen of a multi-lumen catheter is by convention referred to by a nominal gauge (G) derived from the observed flow rate. It should be noted that American wire gauge (AWG) is not the same as SWG.

Table App. 5 French and gauge cross reference (1 inch = 25.4 mm)

FRENCH	INCHES	mm	GAUGE (SWG)
	0.016	0.477	27
	0.018	0.457	26
	0.020	0.508	25
	0.022	0.559	24
	0.024	0.610	23
	0.028	0.711	22
	0.032	0.813	21
	0.035	0.889	20
3	0.039	0.975	
	0.042	1.067	19
	0.049	1.245	18
4	0.053	1.346	
	0.058	1.473	17
	0.065	1.651	16
5	0.066	1.676	
	0.072	1.829	15
6	0.079	2.001	
	0.083	2.108	14
7	0.092	2.337	
	0.095	2.413	13
8	0.0105	2.667	
	0.109	2.769	12
9	0.118	2.997	
	0.120	3.048	11
10	0.131	3.327	
	0.134	3.404	10
11	0.144	3.658	
12	0.158	4.013	
13	0.170	4.318	
14	0.184	4.674	
15	0.197	5.004	
16	0.210	5.334	

Table App. 6 Useful conversions

1 metre	= 1.0936 yards = 3.2808 feet = 39.3696 inches
1 kilogram	= 2.2046 pounds = 35.2736 ounces
1 litre	= 0.22 (imperial) gallons = 0.27 (US) gallons = 1.76 pints
1 kilopascal (100 N m^{-2})*	= 0.146 psi = 7.50 mm Hg = 10.20 cm H_2O
1 mmHg**	= 1.36 cm H_2O = 13.33 $N.m^{-2}$ (≡ 0.1333 kPa) = 0.0194 psi
1 psi	= 6.89 kPa = 51.71 mm Hg = 70.33 cm H_2O
1 joule	= 107 ergs = 0.239 calories
0 K	= –273 °Celsius ('absolute zero')
273.15 K	= 0 ° = 32 °F
373.16 K	= 100 °C = 212 °F
K → °C	= –273.15
°C → K	= +273.15
°C	= (°F – 32) × 5/9
°F	= (°C × 5/9) + 32

*See Table App. 2, 1 pascal = 1 N m^{-2}
**This unit may also be referred to as torr.

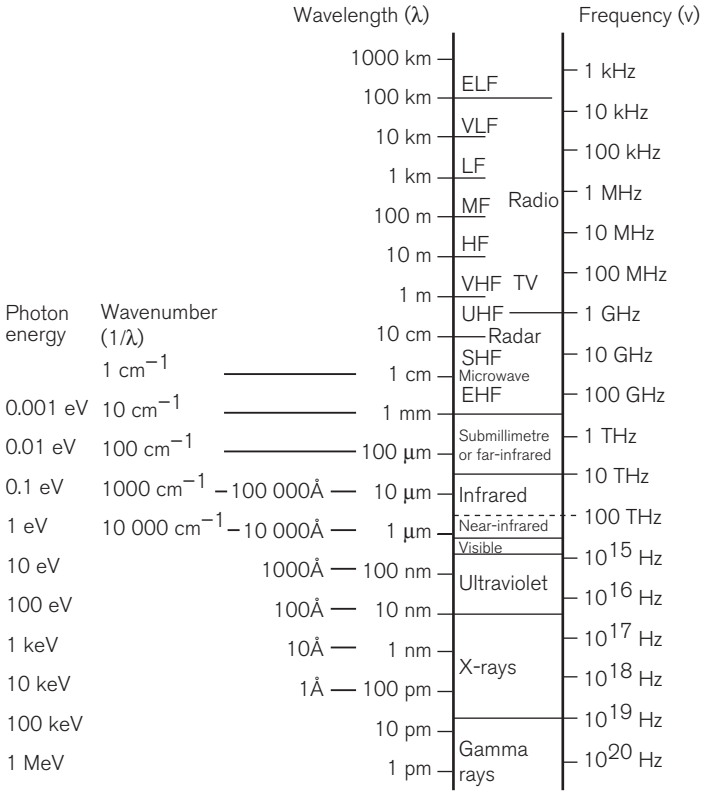


Figure App. 1 The electromagnetic spectrum. eV = electron volt, a unit of energy equal to the work done by an electron accelerated through a potential difference of one volt. Å = angstrom, a unit of length equal to 10^{-10} m. The ground state diameter of a hydrogen atom is about 1 Å.

FURTHER READING

1. http://physics.nist.gov/cuu/Units/prefixes.html (09/10/11)

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