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

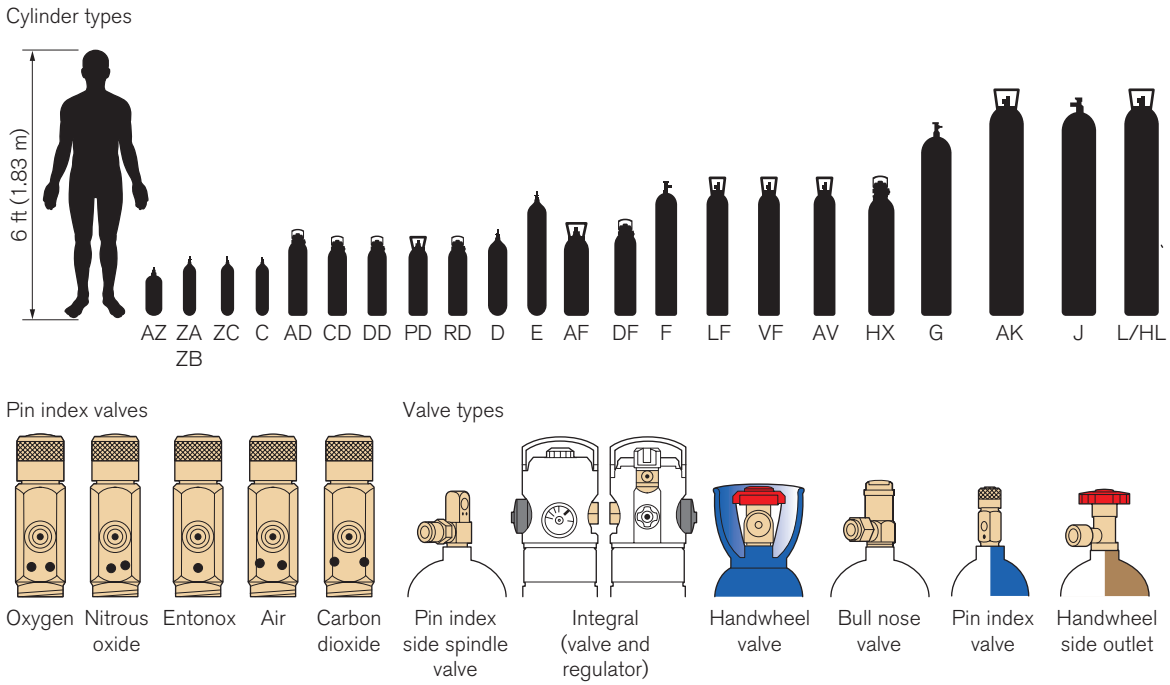


Figure 1.1 Cylinder types and sizes, valves types and pin index valves. *(from BOC Medical, with permission).*

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.

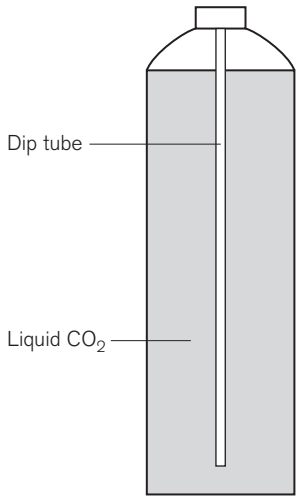


Figure 1.2 LF type CO_2 cylinder with dip tube for delivering liquid CO_2.

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.

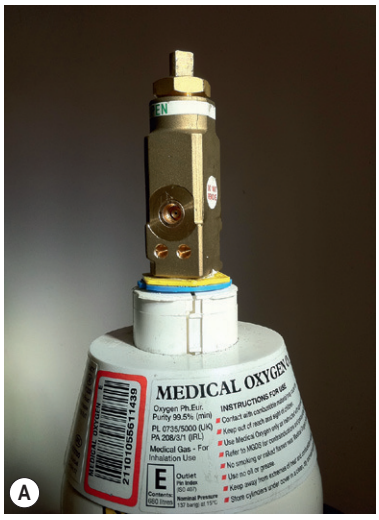
Pin index cylinders require a washer (seal) between the face of the cylinder valve outlet and the equipment to which it is fitted. This bonded non-combustible seal, known as a 'Bodok' washer (Figs 1.7 and 1.8), must be kept clean and should never become contaminated with oil or grease. If a gas tight 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.



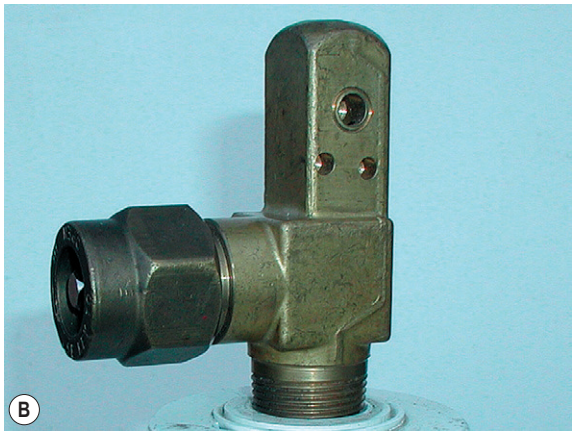


Figure 1.6 **A,** Small pin index valve on E size cylinder. **B,** Side spindle pin index valve.

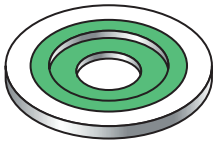


Figure 1.7 Bodok washer.

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.

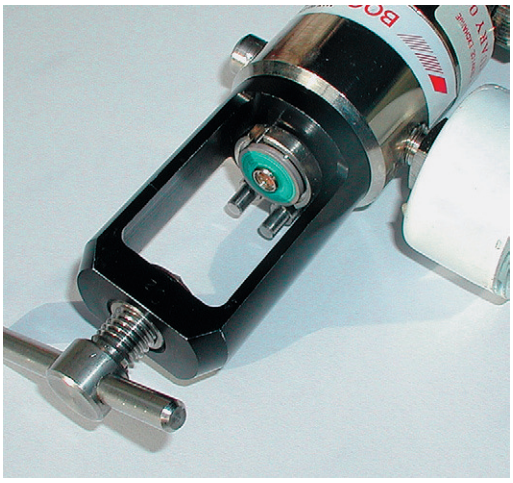


Figure 1.8 Bodok washer fitted to a regulator.

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.

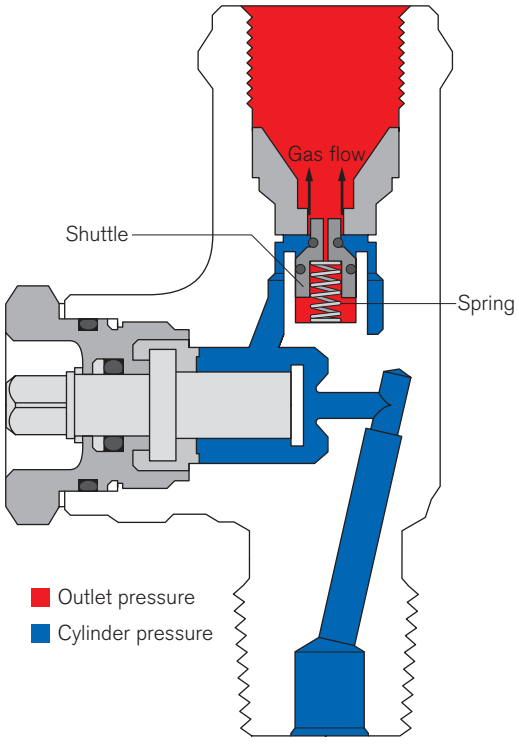


Figure 1.10 Schematic showing 'residual pressure' device of a bull nose cylinder valve.
Redrawn using material kindly provided by Müller Gas Equipment A/S, Denmark.

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.

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

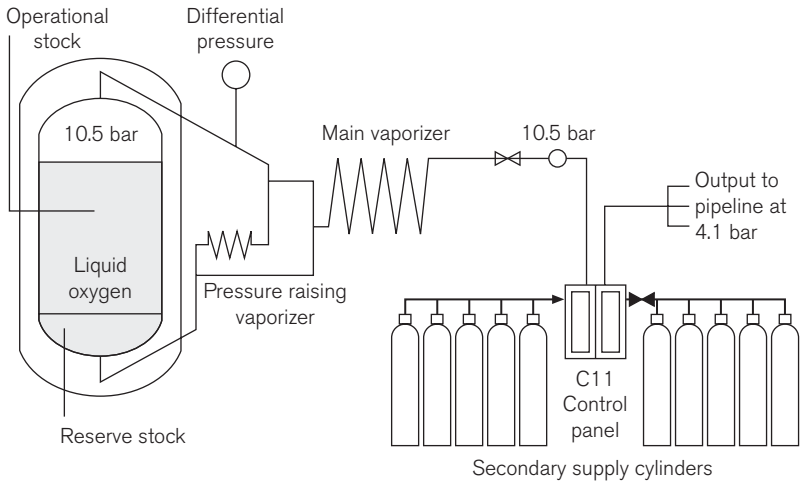


Figure 1.15 Simplified schematic of a single vessel cryogenic liquid system.

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

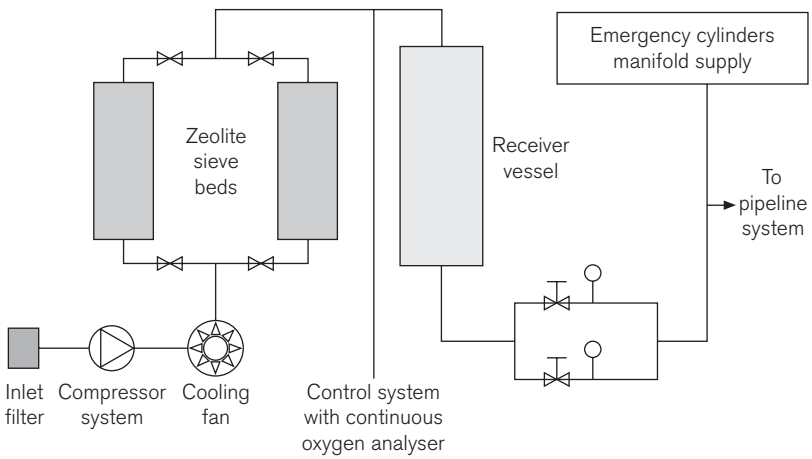


Figure 1.18 A schematic layout for a hospital system.

(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.

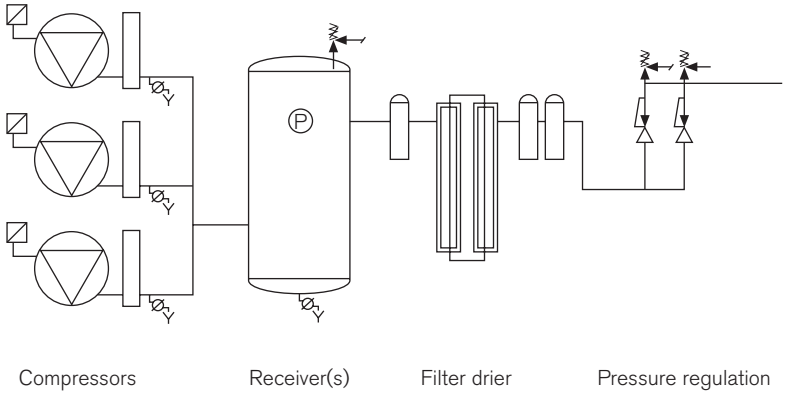


Figure 1.20 Main components of a typical medical compressed air plant.

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

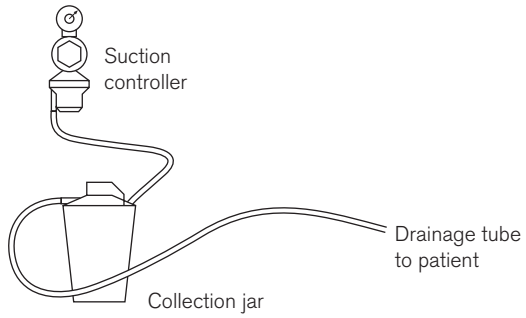


Figure 1.22 Typical components of a ward suction set.

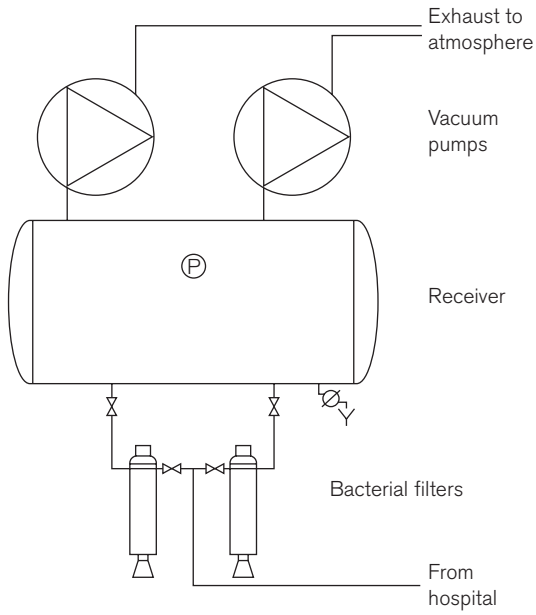


Figure 1.23 Main components of a typical medical vacuum plant.

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.

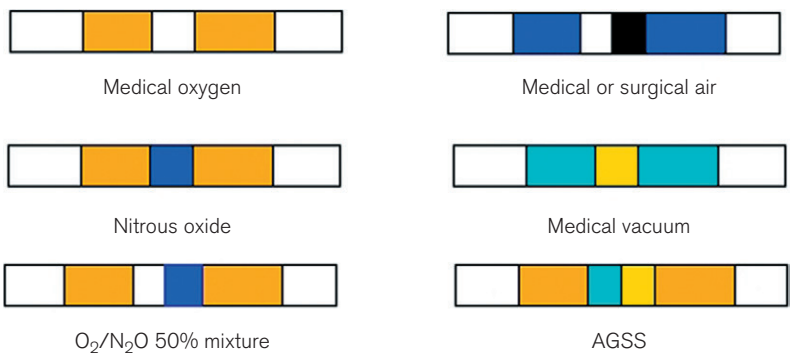


Figure 1.24 Colour codes for medical gas pipework.

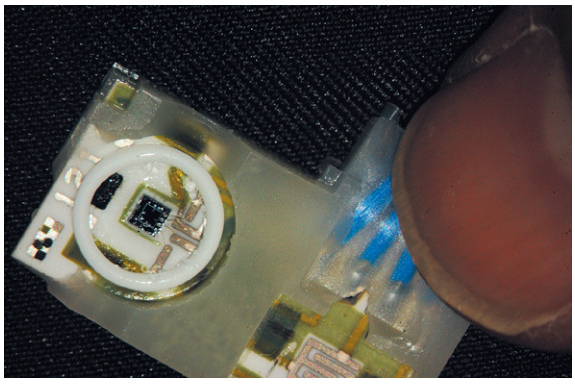


Figure 2.8 A solid-state electronic pressure transducer as used in a typical single-patient use invasive arterial pressure monitoring set. The coupling gel or 'membrane', which is interposed between the lumen of the monitoring line and the transduction chamber, has been removed; remnants of this remain 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.

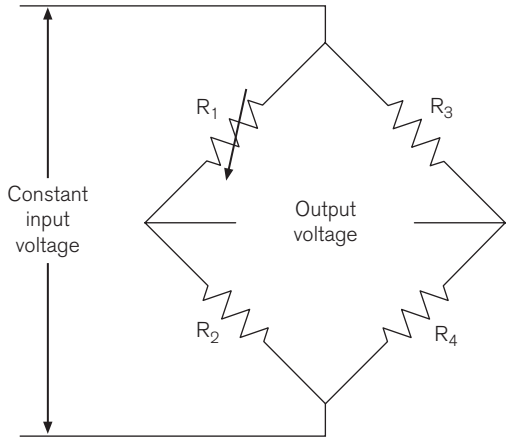


Figure 2.9 A Wheatstone bridge circuit. In the Wheatstone bridge the output voltage is balanced and equal to zero when $R_1/R_2 = R_3/R_4$ for a constant input voltage or current.

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.

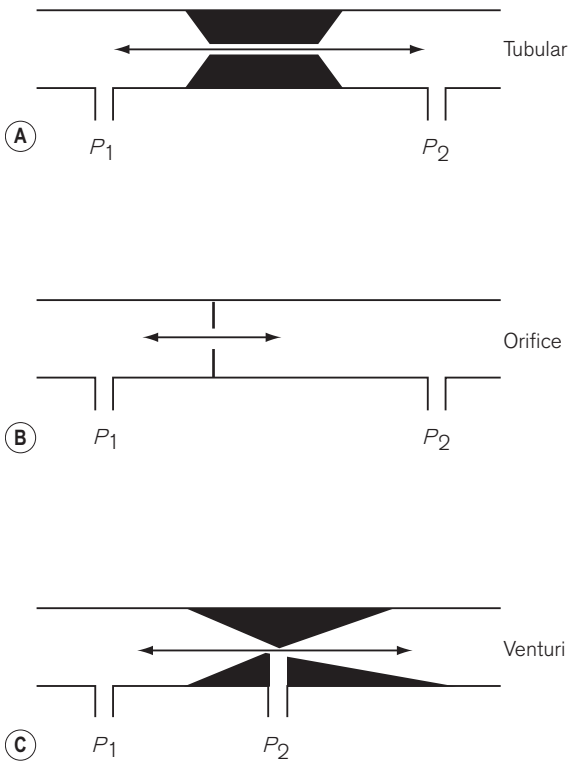


Figure 2.10 Differential pressure flowmeters. P_1, P_2, differential pressures.

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.

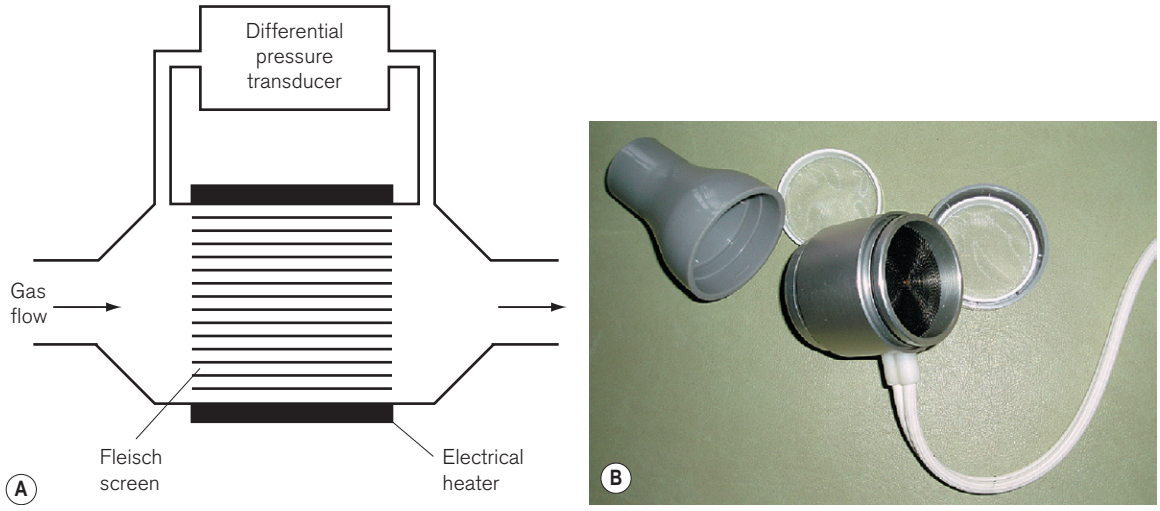


Figure 2.11 Fleisch pneumotachograph. **A.** Schematic and **B.** clinical example.

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.

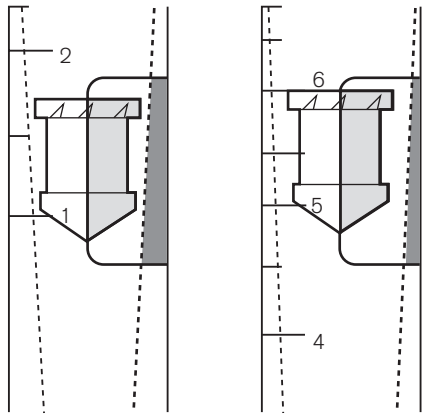


Figure 2.12 The rotameter. In each case, a portion of the tube has been cut away to show that the gap, or annulus, varies with the flow rate. The calibration should be read from the top of the float (e.g. in the right-hand diagram the flow rate is 6 l min^{-1}).

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

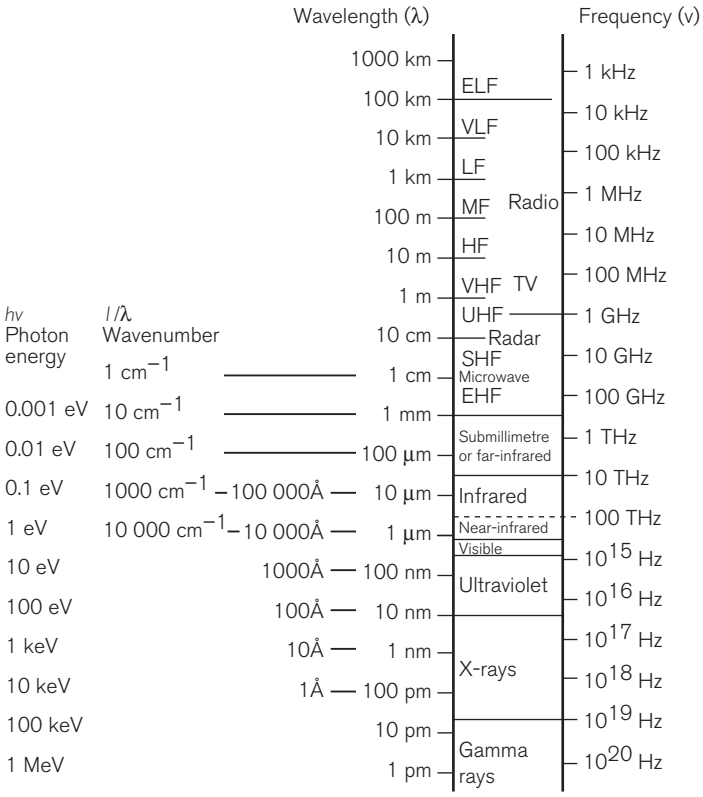


Figure 2.13 A flowmeter tube with varying taper to give an elongated scale at lower flow rates but allowing calibration for high flow rates also.

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

Chapter 3

Vaporizers

Andrew J Davey

CHAPTER CONTENTS

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.

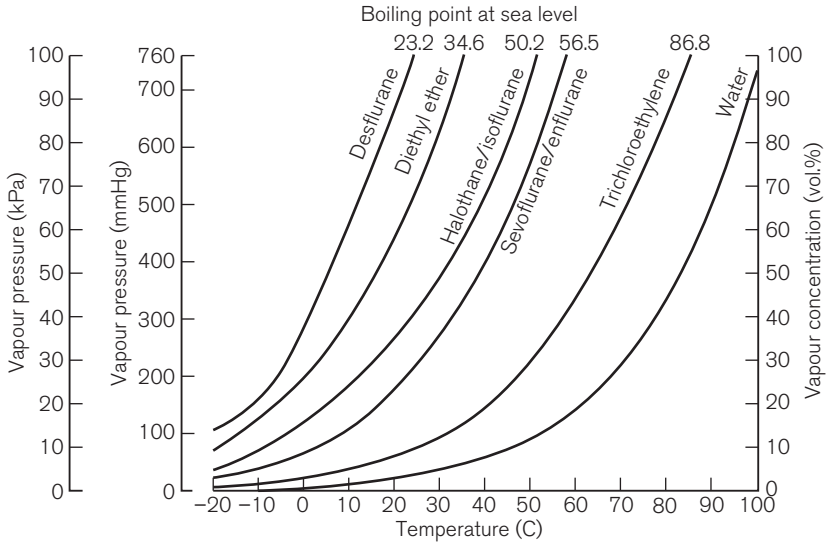


Figure 3.1 Vapour pressure curves for anaesthetic agents.

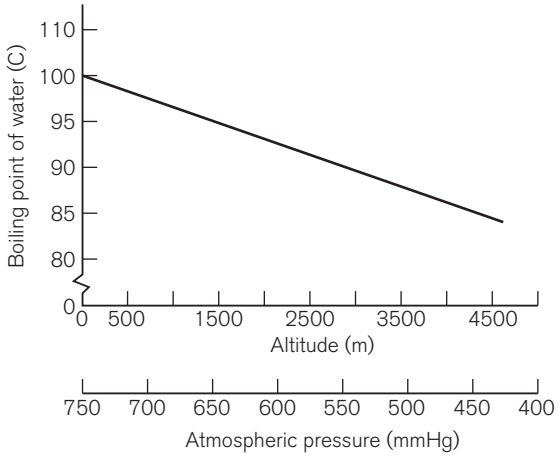


Figure 3.2 Variation of boiling point of water with atmospheric pressure or altitude.

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*.

A modern vaporizer needs to be constructed so that it provides a suitable, stable and predictable concentration of anaesthetic vapour for admixture with other patient gasses. The delivered amount of vapour must not be affected by changes in temperature, and flow rates of other gasses.

TYPES OF VAPORIZER

Appropriate vaporization may be achieved by either:

- splitting the patient gas flow so that only a portion passes through the vaporizer. This picks up saturated vapour and then leaves to mix with the remainder of the gas that has gone through a bypass. The final concentration may be altered by varying the splitting ratio between bypass gas flow and vaporizer gas flow, using an adjustable valve. This type is often referred to as a *variable bypass vaporizer* (Fig. 3.3); or
- alternatively, the vaporizer can be constructed so that it heats the anaesthetic agent to a temperature above its boiling point (in order that it may behave as a gas) and which can then be metered into the fresh gas flow (Fig. 3.4A). Similarly, a vaporizer may contain a fine metal sieve that is submerged in the anaesthetic agent and through which a small independent and metered gas supply (normally oxygen) can be made to pass. The minute bubbles produced have a very large surface area and produce a saturated vapour at ambient pressure, which can then be passed into the fresh gas flow (Fig. 3.4B). These types of vaporizer are often referred to as *measured flow vaporizers*.

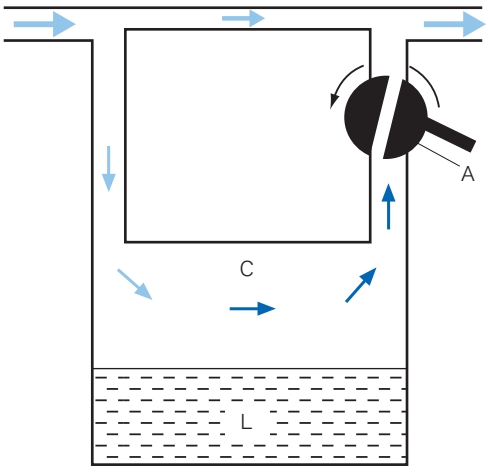


Figure 3.3 A schematic diagram of a variable bypass vaporizer. A flow splitting valve that can be rotated to alter the relative diameters of the vaporizer and bypass channels and so vary the flows through them. L, liquid; C, vaporizer chamber.

It should also be noted that the various anaesthetic inhalational agents currently available have widely differing potencies and physical properties and hence require devices constructed specifically for each agent. Very potent agents (halothane, enflurane, sevoflurane, isoflurane and desflurane) require vaporizers that can accurately control the concentration of vapour leaving the vaporizer. However, agents such as diethyl ether, with a lower potency, may be used safely with simpler apparatus (if necessary), in which the vapour concentration is not accurately known, since there is less risk of over-dosage (see Chapter 27).

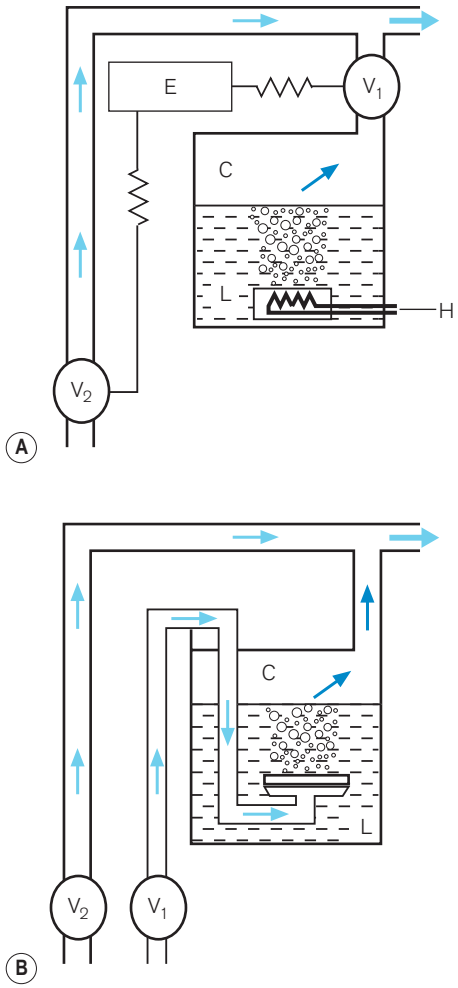


Figure 3.4 Schematic diagrams of measured flow vaporizers. **A.** H, electric heater; L, liquid anaesthetic agent; C, vaporizer chamber; V_1, V_2, flow control valves; E, electronic valve controller to proportion flows. **B.** L, liquid anaesthetic agent; C, vaporizer chamber; V_1, V_2, flow control valves.

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

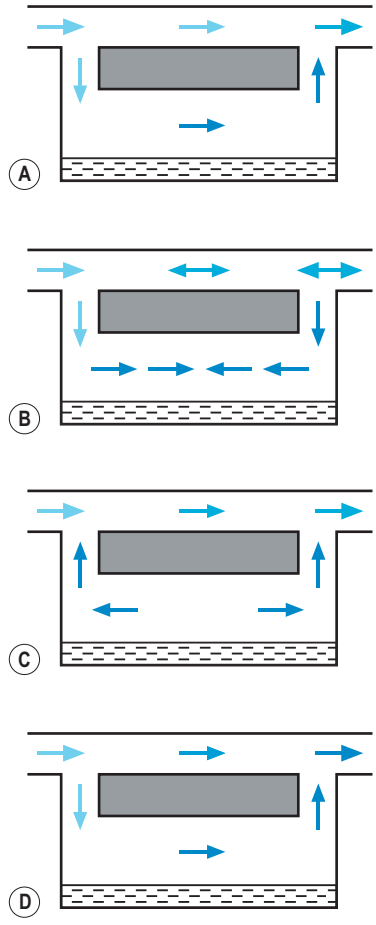


Figure 3.7 **A.** Flow-through vaporizer with no back pressure. **B.** Back-up pressure and/or reverse flow causing build-up of carrier gas in the vaporizer chamber. **C.** Release of back pressure causing gas and saturated vapour to escape through the vaporizer inlet and outlet and into the bypass.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

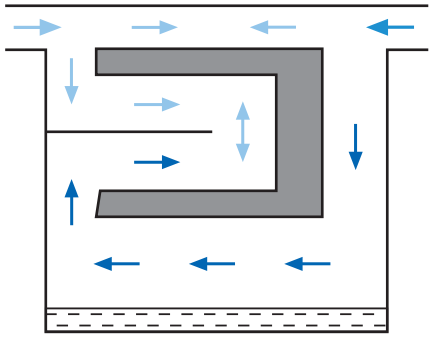


Figure 3.8 Elongation of the inflow channel in a vaporizer preventing saturated vapour reaching the bypass.

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.

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.

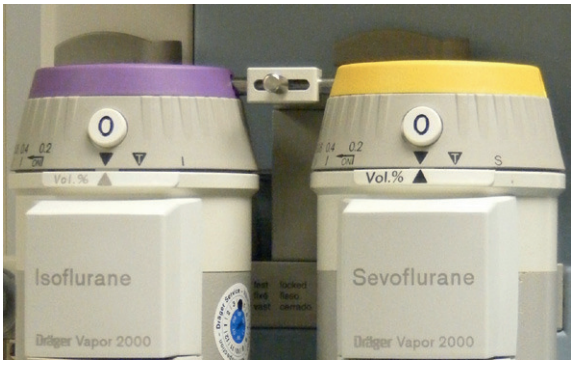


Figure 4.22 Dräger Interlock 2 system, and Vapor 2000 vaporizers.

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.

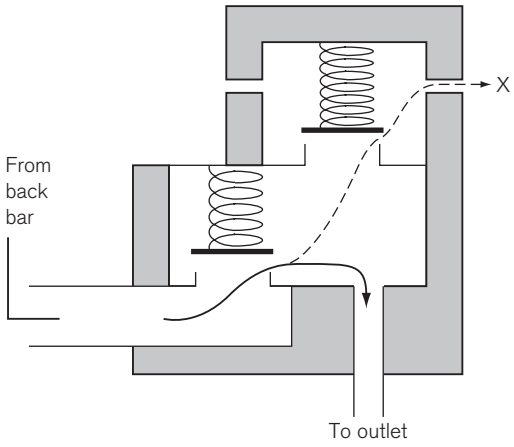


Figure 4.23 Combined non-return and pressure-relief valve at the end of the back bar. If the outlet is obstructed, the gasses escape at X, so protecting the back bar from overpressure. A low-pressure relief valve is also available to protect the patient.

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.

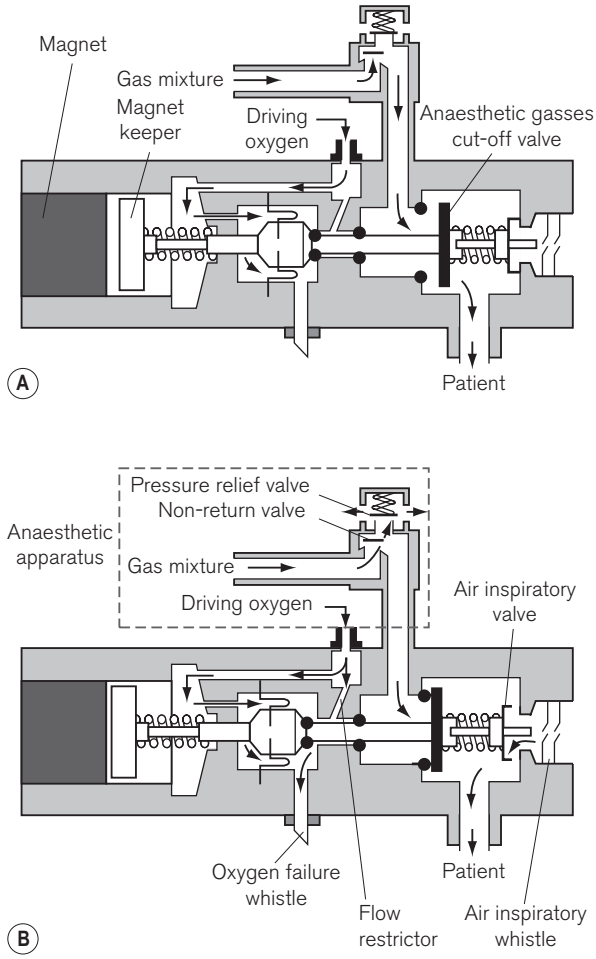


Figure 4.24 Oxygen failure warning device. **A.** Normal operation. **B.** Operation during oxygen failure.

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.

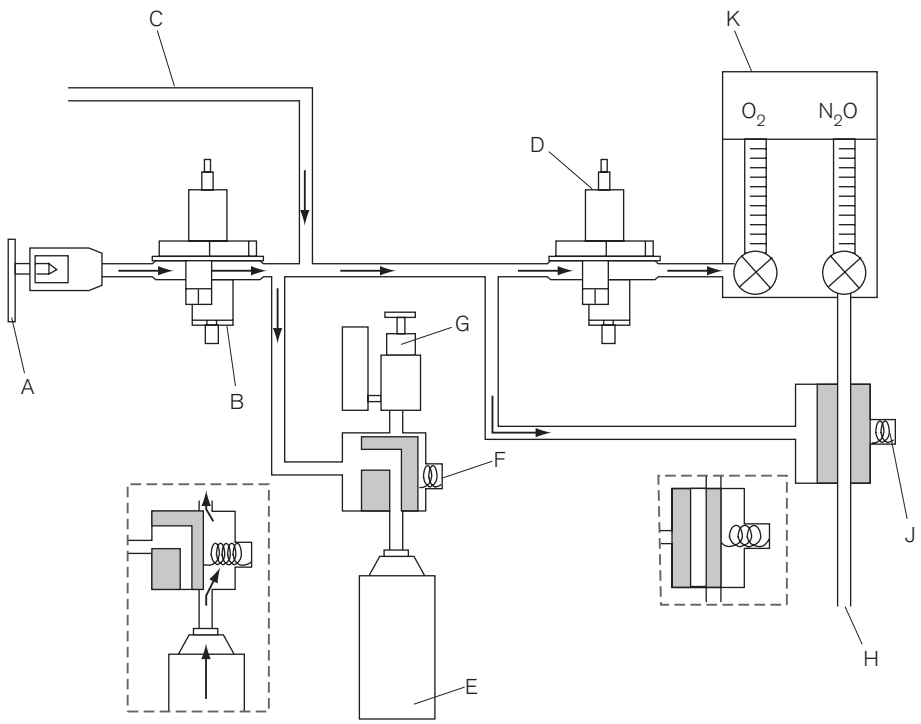


Figure 4.25 A schematic diagram of a current oxygen failure warning device. A, Cylinder yoke for oxygen; B, primary regulator for oxygen (13 700 kPa to 420 kPa); C, pipeline oxygen supply; D, secondary regulator for oxygen (420 kPa to 140 kPa); E, reservoir of oxygen required to power the Ritchie whistle 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.

Common gas outlet

The various medical gasses and vapours exit the machine via a coaxial 22 mm male/15 mm female conically tapered outlet. Machines must have no more than one common gas outlet, although if there is an integral circle breathing system, gas flow may be switched between this and the common gas outlet (CGO).

This CGO may be fixed, or swivelled through 90° (Cardiff Swivel), and should be strong enough to withstand a bending moment of force of up to 10 Nm applied to its axis, since heavy equipment is often attached.

Previously the CGO was often designed with a metal disk in its base to act as a one-way valve and allow entrainment of room air through a high-pitched inspiratory whistle in situations where gas flow was insufficient to meet patient demands during spontaneous ventilation. This facility is now rarely installed as these tilt valves were prone to leaking from impaction of grit on the valve seating. Furthermore, the increased popularity of circle breathing systems for which such a valve serves no function has rendered them obsolete.

Auxiliary gas sockets

Anaesthetic machines may now be fitted with mini-Schrader gas sockets (Fig. 4.26), but only for air or oxygen. These may be used to power devices such as ventilators, gas injection systems for bronchoscopy, and suction units. The sockets should be permanently and legibly marked for their specific gasses (air or oxygen) and their working pressure of 400 kPa approximately (in the UK).

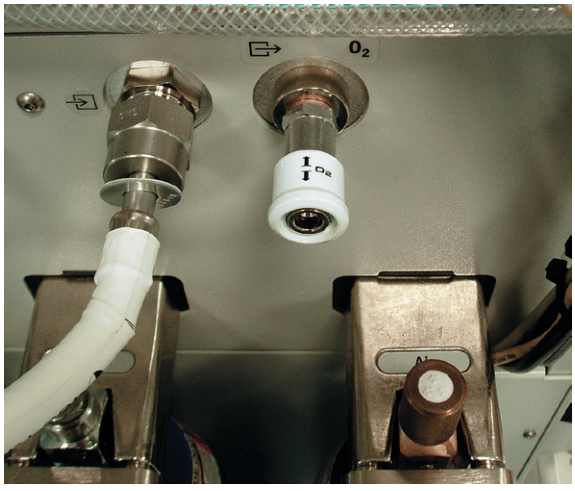


Figure 4.26 An auxiliary oxygen outlet on the back of a Datex-Ohmeda Aestiva/5 anaesthetic workstation.

VENTILATORS

Automatic ventilators, including their classification and fundamental principles, are discussed specifically in Chapter 9. Ventilators for use in the Intensive Care Unit (ICU) are further considered in Chapter 10.

From the point of view of the anaesthetic workstation, ventilators have developed significantly even in the last decade. Whereas once-simple mechanical minute volume dividers placed on the anaesthetic machine were the norm, with a few adventurous manufacturers installing pneumatically driven bag squeezers or intermittent blowers, the modern anaesthetic ventilator is now akin to an ICU ventilator.

Over the past few years, software driven ventilators have become standard, with manufacturers able to offer a variety of ventilatory modes including pressure-controlled ventilation, synchronized ventilation and pressure-supported spontaneous ventilation as optional extras with the machine's in-built ventilator. This flexibility is largely a function of the sophisticated electronic control of the proportional flow valve (see below).

In order to fully separate the gasses of the pneumatically driven ventilator used by most manufacturers (there are exceptions such as the Dräger Primus and Zeus below) from the respiratory gasses of the patient, the *enclosed bellows* arrangement with a rising bellows is almost ubiquitous. Chapter 9 gives a detailed overview of ventilators with specific reference to the 'bag squeezer' type ventilators seen in modern anaesthetic workstations. The Blease900 Series ventilator (Spacelabs Healthcare) is described in detail in that chapter and gives an insight into the modern ventilators used in many anaesthesia machines.

Some of the hardware for such ventilators is briefly described in the paragraphs that follow here. Significant variations in design of anaesthesia workstation ventilator are described in the latter part of this chapter under the individual machines, in order to give a comprehensive picture of the devices available.

Proportional flow valves

The heart of the modern ventilator is the proportional flow valve. Proportional valves are increasingly also seen in the control of gas flow within the patient gas circuits. This type of valve allows extremely accurate and rapid control of the flow of gas from a high-pressure source. Whereas a simple solenoid valve can oscillate rapidly between on and off positions, the proportional valve is so termed because the excursion of the solenoid (and opening of the attached valve) is proportional to the voltage applied across it (Fig. 4.27). Fig. 4.28 shows a typical proportional valve. Such a valve typically allows flow rates up to 130 l min^{-1}

Anaesthesia, 1996, Volume 51, pages 939–942

APPARATUS

Anaesthesia with the Boyle's bottle vaporizer

S. A. WHITE AND L. STRUNIN

Summary

Having used the Boyle's bottle vaporizer apparatus out of necessity in a developing country, the concentration of agent that had been administered was investigated retrospectively. Three anaesthetic agents, halothane, isoflurane and enflurane, were measured at different temperatures, using a Boyle's anaesthetic machine and a Boyle's bottle in circuit with a Magill breathing system connected to a Rascal II Agent Monitor. Bubbling a fresh gas flow of 5 l.min^{-1} through the anaesthetic liquids generated concentrations in excess of 12%. Elevating the initial temperature of the vaporizer increased the delivered concentration, although this effect was short-lived. Therapeutic concentrations of vapour were achieved for all three agents by avoiding bubbling and manipulating the 'splitting ratio' lever. The Boyle's bottle vaporizer may be used with modern anaesthetic agents such as halothane, isoflurane and enflurane. However, the limitations of and variations between vaporizers should be borne in mind. An agent monitor employed at the patient end of the circuit would be an important safety feature.

Key words

Anaesthetics, volatile; halothane, isoflurane, enflurane.
Equipment; Boyle's bottle.

On a recent trip with Operation Smile International to Kenya, short anaesthetics were required for plaster cast changes in children who had undergone club foot repairs. To avoid delaying surgery in the operating theatres, these procedures were performed in the theatre recovery room. This was equipped with an anaesthesia machine (Boyle) comprising oxygen and nitrous oxide flowmeters, a Fluotec Mark III halothane vaporizer and a trichlorethylene Boyle's bottle vaporizer. A check of the anaesthesia machine revealed that the Fluotec vaporizer was stuck in the 'off' position and was unusable. No other machine was available and therefore it was decided to use the Boyle's bottle vaporizer with halothane for anaesthesia. As only one of our anaesthetists had used such a vaporizer before, a quick tutorial was instituted.

The vaporizer is an uncalibrated, temperature-uncompensated plenum type. To turn the vaporizer on, the lever (Fig. 1(a)) is progressively rotated through an uncalibrated arc with four arbitrary marks: 'OFF 1, 2 and 3'. In the latter position, the fresh gas flow is totally diverted through the vaporizer. Any position less than '3' represents a crude 'splitting ratio' with some fresh gas flow bypassing the vaporizer. However, the split varied considerably between vaporizers as it was not based on any of the calculations used for modern concentration-calibrated vaporizers. The surface area available for vaporisation can be dramatically increased by lowering the plunger (Fig. 1(b)) and thus bubbling the fresh gas flow through the anaesthetic liquid [1]. The vaporizer was originally designed for diethyl ether, but was used subsequently for trichlorethylene, halothane and methoxyflurane [1–3]. At the time, it was stressed that the fresh gas flow should never be 'bubbled' for these latter three agents, as this might lead to uncontrolled high concentrations. Indeed, after the introduction of halothane, Boyle's bottles were modified by removing the plunger [4] thus preventing 'bubbling'. Subsequently, the modern calibrated, temperature-compensated vaporizers, with a wick or baffle system to increase surface area for evaporation were developed.

In Kenya, 10 unpremedicated children (aged 4–10 years) were anaesthetised uneventfully with oxygen, nitrous oxide (30:70% mixture) and halothane by inhalation. Pulse oximetry was used for monitoring and although some slowing of heart rate was observed, there were no other untoward effects. All the anaesthetists in the team were impressed with the rapidity of inhalational induction of anaesthesia using a Magill system, with the Boyle's bottle vaporizer full on and the plunger just above the liquid. This suggested that the concentration of halothane was high. With the plunger up and the lever in the '2' position,

S.A. White, MB, BS, FRCA, Clinical Research Fellow, L. Strunin, MD, FRCA, FRCPC, BOC Professor of Anaesthesia, Anaesthetics Unit, The Royal London Hospital, 4th Floor Alexandra Wing, Whitechapel, London E1 1BB.
Correspondence should be addressed to Professor L. Strunin.
Accepted 9 March 1996.

0003 2409/96/010939 + 04 $18.00/0

adequate anaesthesia with spontaneous ventilation was maintained for the duration of the procedures. Recovery was not unduly prolonged.

On returning to the UK, we set out to determine, under laboratory conditions, the concentration of anaesthetic agent generated by this apparatus.

Methods

In order to determine the concentration of volatile agents delivered by a Boyle's bottle, an anaesthesia machine (Fig. 2) similar to the one used in Kenya was mobilised from the anaesthetic museum at the Royal London Hospital. Of interest is the Bosun's oxygen failure alarm next to the flowmeter bank. The latter contains Rotameters for carbon dioxide and cyclopropane, in addition to oxygen on the left and nitrous oxide on the right side. Oxygen and nitrous oxide in a combined flow of 5 $l.min^{-1}$ was set, the Boyle's bottle was filled with 100 ml of anaesthetic agent and a Magill system was connected to the common gas outlet. At the 'patient' end of the circuit, gases were sampled continuously by a Rascal II gas analyser [5, 6]. The Rascal uses the principle of Raman scattering of laser light to identify and measure halothane, enflurane and isoflurane concentrations in a gas mixture. The instrument compares favourably with conventional infrared [7], photo-acoustic [8] and mass spectrometry [5] measurement of volatile anaesthetics.

Results

Table 1 shows the concentration of halothane, enflurane and isoflurane at various settings of the vaporizer. This was generated from the original tracings produced from the Rascal II monitor (Fig. 3) during the experiment. It can be seen from Table 1 that there was an abrupt increase in concentration with all three agents when the fresh gas flow was bubbled through the liquid agent in the Boyle's bottle and the concentrations generated exceeded 12%. These were maintained at greater than 6% for at least 3 min, after which the concentration tended to plateau at a lower concentration. Table 2 demonstrates that the maintenance concentration generated by the bottle settings used in the field was 0.6% halothane.

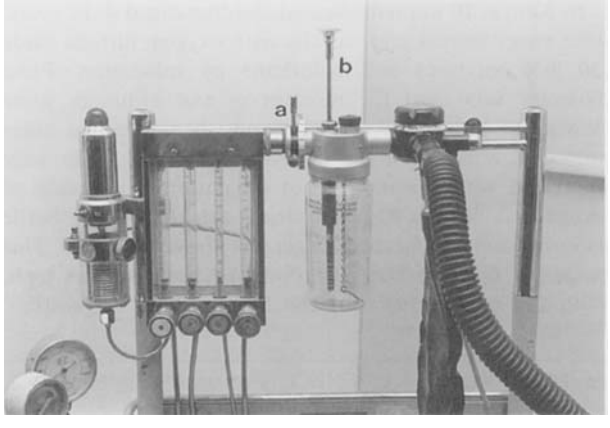


Fig. 1. Boyle's bottle vaporiser: (a) 'Splitting ratio' lever in position '3', (b) Plunger.

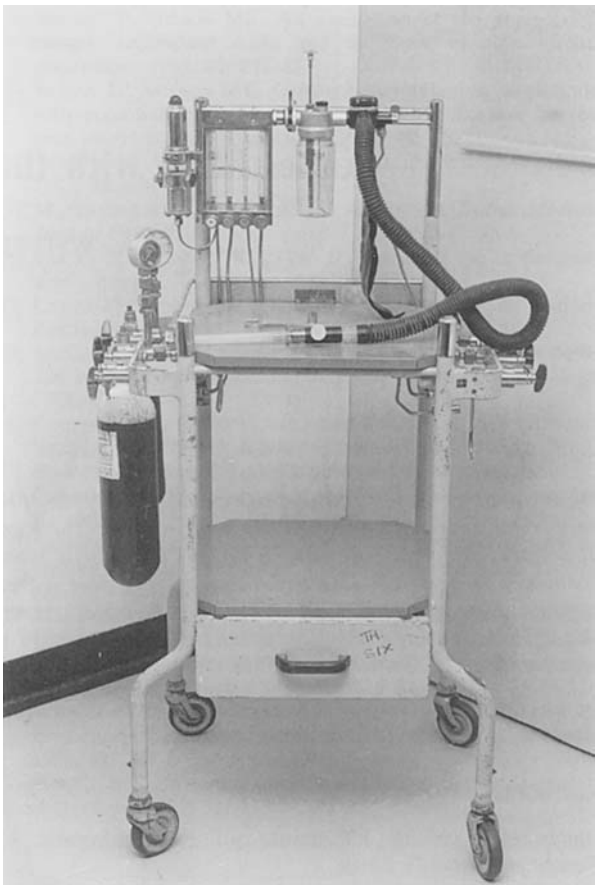


Fig. 2. Boyle's Anaesthesia machine.

However, Nairobi is at an altitude of about 5000 feet, where the atmospheric pressure is 85 kPa (646 mmHg). Since vapour pressure is independent of atmospheric pressure, the theoretical maximum concentration of halothane achievable in Nairobi would be 38%, in contrast to 32% at sea level assuming an ambient temperature of 20°C. The ambient temperature in Nairobi is higher than London potentially producing even higher peak and maintenance concentrations. In our experimental model, this temperature difference was simulated by a 5°C increase in the starting temperature of the liquid agent. This was achieved by initially placing the Boyle's bottle in a temperature-controlled water bath. The water bath was removed and the vaporizer turned on. For halothane, the starting concentration was 17.8% at a starting temperature of 25°C and 14.5% at 20°C. However, the output of the warmed vaporiser decreased rapidly as the ambient temperature was 20°C and the vaporizer is temperature non-compensated.

Table 2 also demonstrates an interesting paradox. Once the splitting ratio has been reduced by moving the lever to position '1' or '2', submerging the plunger reduces the agent concentration substantially. This is because the proportion of gas diverted into the bottle chamber cannot generate sufficient driving pressure to bubble through the liquid. Therefore, an excessively high concentration of volatile agent can only occur if the fresh gas flow is fully diverted into the bottle when the plunger is in the down position and 'bubbling' occurs.

Table 1. The maximum concentration of each agent achieved at a fresh gas flow (FGF) of 5 l.min^{-1}, with the control lever in the '3' position producing full diversion of the fresh gas flow into the bottle chamber. The plunger elevation was the only variable. The theoretical maximum concentration of the agent is expressed assuming equilibration of the agent to its saturated vapour pressure at 20°C. %TM, percentage of the theoretical maximum obtainable; %A, actual percentage measured by the Rascal II.

			Plunger position					
			Submerged		Halfway		Up	
FGF; l.min^{-1}	Agent	SVP; kPa	%A	%TM	%A	%TM	%A	%TM
5	Halothane	32	14.0	44	6.0	19	4.2	13
5	Isoflurane	32	16.5	50	6.5	19	3.8	13
5	Enflurane	24	12.5	52	5.0	20	2.7	13

Discussion

The Boyle's bottle vaporizer, used with diethyl ether, was relatively safe as the boiling point of ether is 35°C and the rapid fall in temperature once vaporisation begins, causes the concentration to decrease quickly, even when 'bubbling'. It is often assumed this is not the case with the modern agents as most have much higher boiling points. However, if the only difference between ether and halothane, for instance, was their boiling points, then ether vaporisation would result in much higher concentrations than those seen with halothane. However, there are also significant differences in the latent heat of vaporisation, saturated vapour pressure and minimum alveolar concentration of the two agents. Of these, the latent heat of vaporisation and minimum alveolar concentration are considerably lower for halothane than ether. As a result, halothane in an uncalibrated vaporizer leads to potentially toxic concentrations.

As can be seen from Table 2, there is an abrupt increase in the concentration delivered by the apparatus on moving the lever position from '2' to '3', reflecting the non-linear relationship of the 'splitting ratio' to the lever position. The lever movement mechanically effects an increase in the size of the aperture allowing the fresh gas flow into the vaporizer. The apparatus design dictates that the increase in aperture bears no relation to the distance that the lever moves and is arbitrary. However, flow along a cylinder (under ideal conditions and ignoring the effects of turbulence) is proportional to the fourth power of the radius, so that one would expect a large increase in flow through the Boyle's bottle for a modest increase in aperture size, as is demonstrated.

In the experimental set-up described above, the high

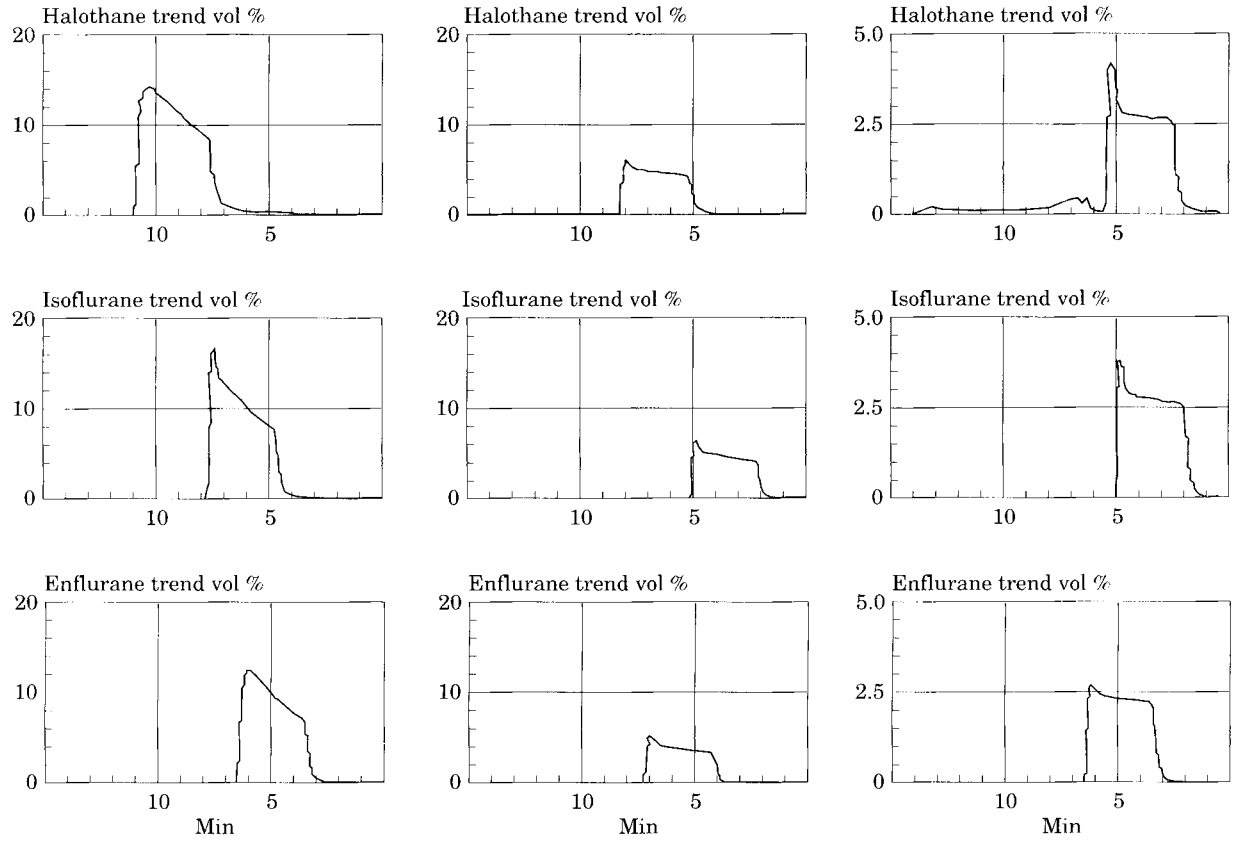


Fig. 3. Rascall II monitor read-outs. Halothane, isoflurane and enflurane as volumes % (*y*-axis) against time in minutes (*x*-axis). The graphs in each segment are with the plunger submerged, halfway up and fully up, as in Table 1.

Table 2. The maximum measured concentrations of halothane generated at a fresh gas flow of 5 l.min^{-1}, by varying the simple controls on the Boyle's bottle. '2'–'OFF', progressive diminution of the 'splitting ratio'; '3', full fresh gas flow diversion to the bottle chamber.

	Lever position			
	'3'	'2'	'1'	'OFF'
Plunger up	6.0%	0.6%	0.6%	0.0%
Plunger submerged	14.5%	0.1%	0.1%	0.0%

concentrations were maintained above 6% for all agents for greater than 3 min with the plunger submerged. Theoretically, the vaporizer can deliver the saturated vapour pressure of a volatile agent, since, with the lever in the '3' position, all the fresh gas flow is diverted into the vaporizer. However, despite producing high concentrations, the vaporizer does not produce values anywhere near the theoretical maximums. This is probably due to inadequate equilibration within the bottle secondary to thermal limitations and mixing of the effluent 'bottle' gas with that already present in the distal limb of the apparatus at the commencement of gas flow.

The problems of uncontrolled, high concentration delivery by the Boyle's bottle can be exacerbated by inadvertent filling of the J-tube during replenishment of the vaporizer by the top filler [4] and shaking of the vaporizer during transport [9]. The latter may be offset by placing a nylon mesh in the bottle.

In summary, the Boyle's bottle vaporizer may be used (if circumstances warrant) for induction and maintenance of anaesthesia with modern volatile agents. The maximum concentration should be limited by keeping the plunger up. It should be remembered that the device is totally uncalibrated, is affected by ambient temperature and, in addition, there may be considerable variations in the extent of the 'on/off' lever positions between vaporizers [1]. The use of an agent monitor at the patient end of the anaesthetic circuit would make the use of the Boyle's bottle vaporizer much safer and perhaps more practical.

References

[1] Mapleson WW. Trichlorethylene concentration from a 'Boyle' type anaesthetic apparatus. *British Journal of Anaesthesia* 1957; **29:** 3–11.

[2] Hill EF. Percentage of fluothane vapour delivered from a Trilene bottle. *British Journal of Anaesthesia* 1957; **29:** 12–6.

[3] Hoffman JC. Control of concentrations from a Boyle apparatus. *Anesthesiology* 1970; **32:** 372–4.

[4] Jennings AMC, Hersant, ME. Increase in halothane concentration following refilling of certain vaporisers. *British Journal of Anaesthesia* 1965; **37:** 137–8.

[5] Palayiwa E, Grint E, Winyard J, Flynn PJ, Strunin L. An evaluation of the Rascal II monitor. *British Journal of Anaesthesia* 1993; **70:** 485P.

[6] Lockwood GG, Landon MJ, Chakrabarti MK, Whitwam JG. The Ohmeda Rascal II. A new gas analyser for anaesthetic use. *Anaesthesia* 1994; **49:** 44–53.

[7] McPeak HB, Palayiwa E, Madgwick R, Sykes MK. Evaluation of multigas anaesthetic monitor: the Datex Capnomac. *Anaesthesia* 1988; **43:** 1035–41.

[8] McPeak HB, Palayiwa E, Robinson GC, Sykes MK. An evaluation of the Brüel and Kjær monitor 1304. *Anaesthesia* 1992; **47:** 41–7.

[9] Seed RF. Vaporisation of halothane during movement. *Anaesthesia* 1967; **22:** 659–63.

Br J Anaesth. 1983 Nov;55(11):1053-60.

Servo-controlled closed-circuit anaesthesia. A method for the automatic control of anaesthesia produced by a volatile agent in oxygen.

Ross JA, Wloch RT, White DC, Hawes DW.

Abstract

A method of closed-circuit anaesthesia has been developed in which the end-tidal concentration of the volatile anaesthetic agent used is controlled automatically using a closed-loop servo system. End-tidal anaesthetic concentrations, measured by the Engstrom EMMA, were maintained in the closed circuit by direct liquid injection. The system was tested in the laboratory and in clinical use (12 subjects). Accurate estimation of anaesthetic uptake was readily obtained and results for the uptake of halothane agreed closely with those of previous workers. The major sources of error in the method were the result of zero offset in the Engstrom EMMA which in turn was caused by humidity and the intrinsic characteristic of the simple proportional controller used. These errors were easily correctable, and end-tidal halothane concentration could be controlled to within 0.1%. Mean halothane vapour uptake at a constant end-tidal concentration of 0.8% was 114 ml min-1 at 1 min, 36 ml min-1 at 5 min, 29 ml min-1 at 10 min and between 22 and 18 ml min-1 at 20-35 min.

PMID: 6639824 DOI: 10.1093/bja/55.11.1053
[Indexed for MEDLINE] Free full text

The Anesthesia Machine: