Module 4

Breathing systems and their components

INTRODUCTION

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

DEFINITIONS

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

CLASSIFICATION OF BREATHING SYSTEMS

These may be classiied according to function:
• non-rebreathing systems (utilizing non-rebreathing valves)
• systems where some rebreathing of previously exhaled gas is possible, but normally prevented by the low of fresh gas through the system

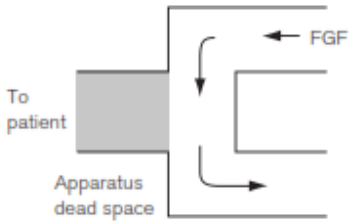

Figure 5.1 Apparatus dead space in a breathing system. FGF, fresh gas flow.

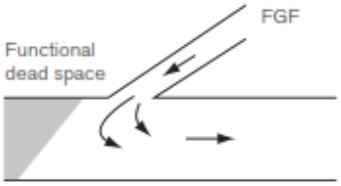

Figure 5.2 Functional dead space in a breathing system with an angled FGF inlet. FGF, fresh gas flow.

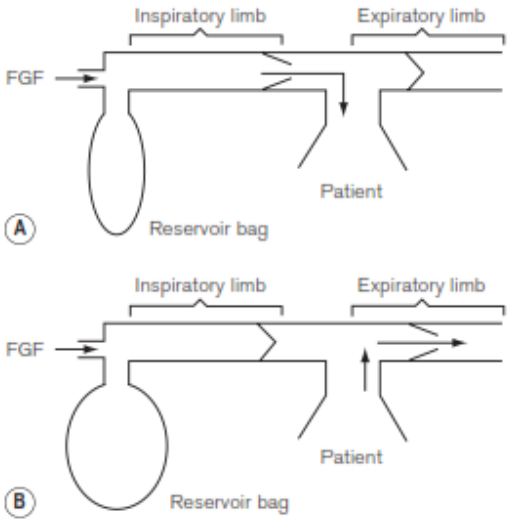

Figure 5.3 Non-rebreathing valve. **A.** Inspiration. **B.** Exhalation. FGF, fresh gas flow.

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

Non-rebreathing systems

- The simplest way to deliver a consistent fresh gas supply to a patient is with a system that utilizes a non rebreathing valve (or valves). Fresh gas enters the system via an inspiratory limb (Fig 5.3A). This is a length of hosing that has a sufficiently wide bore to minimize any resistance to airflow.
- It is reinforced so as to prevent collapse from sub-atmospheric pressure either by manufacturing it with corrugations or by bonding a reinforcing spiral onto the outside of the hose. Both of these methods also allow the tube to bend without kinking. The fresh gas entering is either sucked in by the patient's inspiratory effort or blown in during controlled ventilation and enters the non-rebreathing valve.
- The valve is so constructed that when it opens to admit inspiratory gas, it occludes the expiratory limb Of the system(see Fig. 5.3A). When the patient exhales, the reverse occurs,i.e. the valve mechanism moves to occlude the inspiratory limb and opens the expiratory limb to allow expired gasses to escape (Fig. 5.3B). The inspiratory limb usually includes a bag (1.5–2 litres capacity) that acts as reservoir for fresh gas.

This reservoir contains enough gas to cope with the intermittent high demand that occurs at inspiration. For example, a patient breathing normally (with a minute volume of 7 litres) may well have a tidal volume of 500 ml inhaled over approximately 1 s. This produces an average inspiratory low rate (not peak low rate) of 30 l min Without this reservoir in the system, the fresh gas low rate would have to at least match this igure (probably more, to match the patient's peak inspiratory low rate) in order to avoid respiratory embarrassment.

- The reservoir bag is refilled with fresh gas during the expiratory phase. It can also be compressed manually to provide assisted or controlled respiration since the nonrebreathing valve works equally effectively in this mode as it does for spontaneous respiration. In the non-rebreathing system described, the fresh gas low rate *must not be less than* the minute volume required by the patient.

Systems where rebreathing is possible

A miscellany of breathing systems was developed by early pioneers (largely intuitively) that allowed the to-and-fro movement of inspiratory and expiratory gasses within the breathing system. Carbon dioxide elimination was achieved by the lushing action of fresh gas introduced into this breathing system, rather than by separation of the inspiratory and expiratory gas mixtures by a nonrebreathing valve as described above. As it is mainly the
lushing effect of fresh gas that eliminates carbon dioxide, these systems retain the potential for rebreathing of carbon dioxide when fresh gas low rates are reduced. Mapleson (1954) classiied these systems (A to E) according to their eficiency in eliminating carbon dioxide during spontaneous respiration. An F system, the Jackson Rees modiication of system E (Ayre's T-piece), was later added to the classiication by Willis

Mapleson's classification of breathing systems

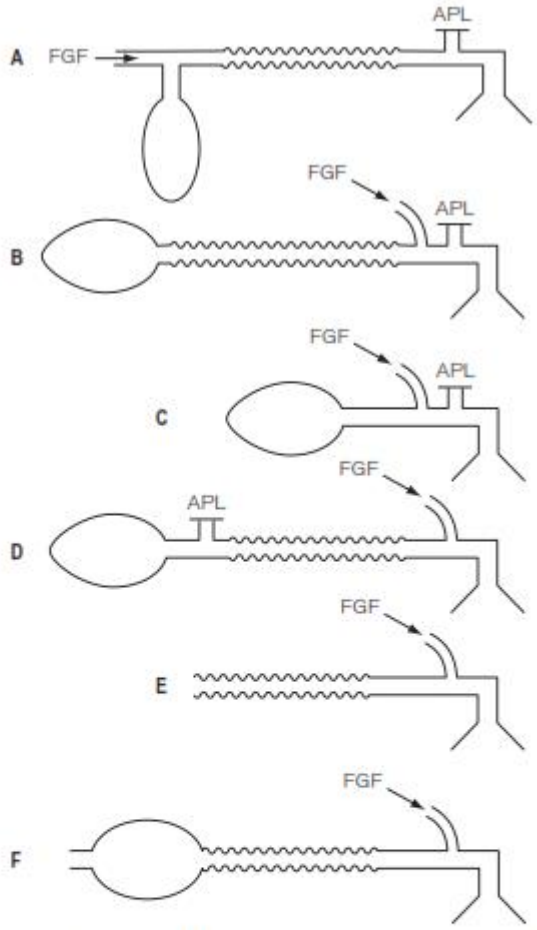

Figure 5.4 A modified Mapleson classification of breathing systems in which there is potential for rebreathing. System **A** houses the gas reservoir in the afferent limb and is alternatively referred to as an afferent reservoir system. The fresh gas flow (FGF) need only be at or just below the patient's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

Fig. 5.4 illustrates a modiied Mapleson classification of breathing systems. These all contain similar components but are assembled in different sequences in order to be used more conveniently in specific circumstances.

However, as a result, the eficiency of each system is different.

They are catalogued in order (A, B, C, D, E, F) of increased requirement of fresh gas low to prevent rebreathing during spontaneous respiration. System A requires.

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0.8–1 times the patient's minute ventilation, B and C require 1.5–2 times the patient's minute ventilation and systems D, E and F (all functionally similar) require 2–3

times the patient's minute ventilation to prevent rebreathing

during spontaneous

respiration.

WORKING PRINCIPLES OF BREATHING SYSTEMS

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

Anatomical dead space gas

Pure alveolar gas

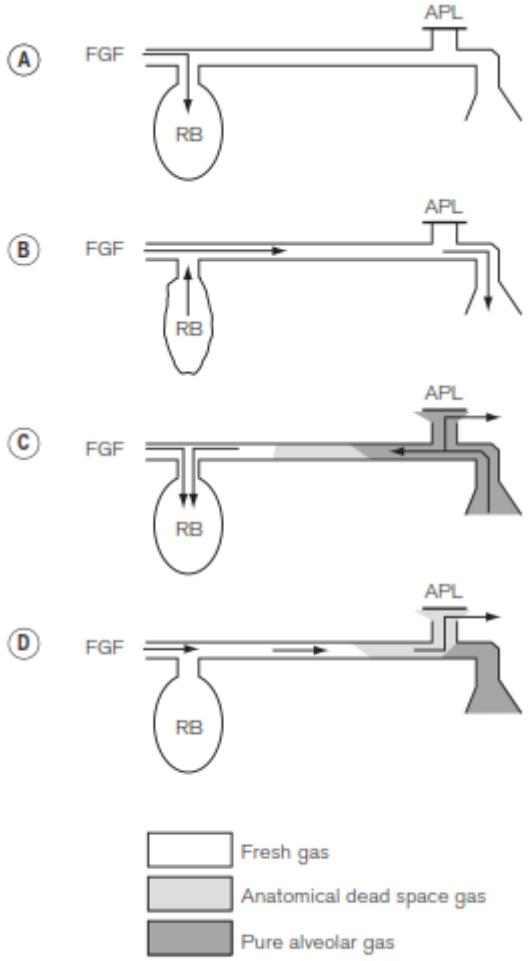


Figure 5.5 The Mapleson A system used with spontaneous breathing (see text). FGF, fresh gas flow; APL, adjustable pressure-limiting valve; RB, reservoir bag.

• At one end, an inlet for fresh gas linked to a 2 litre istensible rubber or neoprene reservoir bag. (Not rebreathing bag, as the patient's exhaled gasses should never be allowed to pass back into it.) This is attached to:

• a length of corrugated breathing hose (minimum length 110 cm with an internal volume of 550 ml).

This represents slightly more than the average tidal volume in an anaesthetized adult breathing spontaneously. This volume is important as itminimizes the backtracking of exhaled alveolar gas

back to the reservoir bag (see below). This is in turn connected to:

• a variable tension, spring-loaded lap valve for venting of exhaled gasses. This valve should be attached at the opposite end of the system from the reservoir bag and as close to the patient as possible.

It will be subsequently referred to as an APL (adjustable pressure limiting) valve.

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

• *The first inspiration.* In Fig. 5.5A the reservoir bag and breathing hose have been illed by an ideal fresh gas low and attached (with a gas-tight it) to the patient, who is about to take a breath. The whole system is therefore full of

fresh gas. As the patient inspires, the gasses are drawn into the lungs at a rate greater than the fresh gas low and so the reservoir bag partially empties as shown in Fig. 5.5B.

• *Expiration.* In Fig. 5.5C the patient has begun to exhale, and because the reservoir bag is not full, the exhaled gasses are breathed back along the corrugated hose, pushing the fresh gasses in the hose back towards the reservoir bag. However, before the exhaled gasses can pass as far as the reservoir bag (hence the importance of the length of the inspiratory hose), the latter has been reilled by the fresh gasses from the corrugated hose plus
the continuing fresh gas low from the anaesthetic machine. A point is reached when the reservoir bag is again full and, as the patient is still exhaling, the remaining exhaled gasses have to pass out through the APL (expiratory) valve, which now opens.

The first portion of exhaled gasses to pass along the corrugated hose from the patient was that occupying the patient's anatomical dead space and, therefore, apart from being warmed and slightly humidiied (a satisfactory state of affairs), is unaltered, not having taken part in respiratory exchange. This is followed by alveolar gas (with a
reduced oxygen content and containing carbon
dioxide), the irst part of which may enter the
corrugated hose, and the rest which is expelled
through the APL valve when the reservoir bag is full.

• *End-expiratory pause.* The next stage is the endexpiratory pause. The fresh gas low entering the system now drives the exhaled gasses, or some That had tracked back along thecorrugated hose,out through the APL valve. It can be seen that the expiratory pause
Is important because it prevents the potential for the rebreathing of exhaled alveolar gasses
That would otherwise be contained in the hose at the end of expiration (Fig.5.5D).

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

patient's minute volume and rebreathing ofalveolar gas is, therefore, prevented. In theory, provided there is no mixing of fresh gas, deadspace gas and alveolar gas and a sufficient end-expiratory pause,the fresh gas low rate need only match alveolar
ventilation (approximately 66% of the minute volume), as in this situation alveolar gas only will be vented through the APL valve.In practice, however, a number of factors dictate a higher fresh gas low rate (70–90%), for instance:

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

Mapleson A system and controlled ventilation

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

- *Inspiratory phase.* The APL valve has to be kept almost closed so that suficient pressure can develop in the system to inlate the lungs. During the first inspiratory phase with the anaesthetist squeezing the bag, some of the fresh gasses are blown out of the valve.

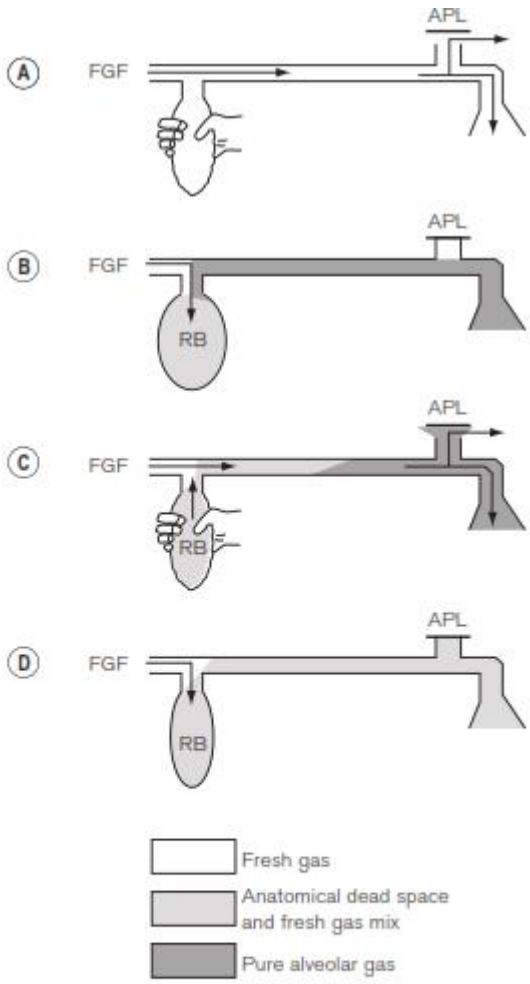


Figure 5.6 Mapleson A system with assisted or controlled ventilation: **A.** at the end of inspiration; **B.** at the end of expiration; **C.** during subsequent inspiration; **D.** at the end of the subsequent inspiration. Note that much rebreathing takes place (see text). FGF, fresh gas flow; APL,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

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

When the anaesthetist squeezes the bag again, the first gasses to enter the patient's lungs will be the previously exhaled alveolar gasses. The volume of gasses escaping via
the APL valve during this second inspiratory phase is initially small (the valve being almost closed), but gradually increases as the pressure in the system rises towards the
maximum inspiration. Therefore, the greatest amount of gas will he dumped late in the cycle and will consist mainly of fresh gas. Under these circumstances there is
considerable rebreathing (Fig. 5.6 see shading). Furthermore, as alveolar gas will have entered the reservoir bag, there will always be carbon dioxide contamination in any subsequent inspirate. In order to prevent this and thereby minimize the potential for rebreathing alveolar gas, a high fresh gas low rate is required. This is usually of the order of 2 times the patient's minute ventilation. This situation is highly wasteful with
regard to fresh gas and also increases the potential for pollution.

Other Mapleson A breathing systems

*The Lack co-axial breathing system(f*ig. 5.7A) The traditional layout of a Magill system sites the APL valve as close to the patient as possible. However, this:

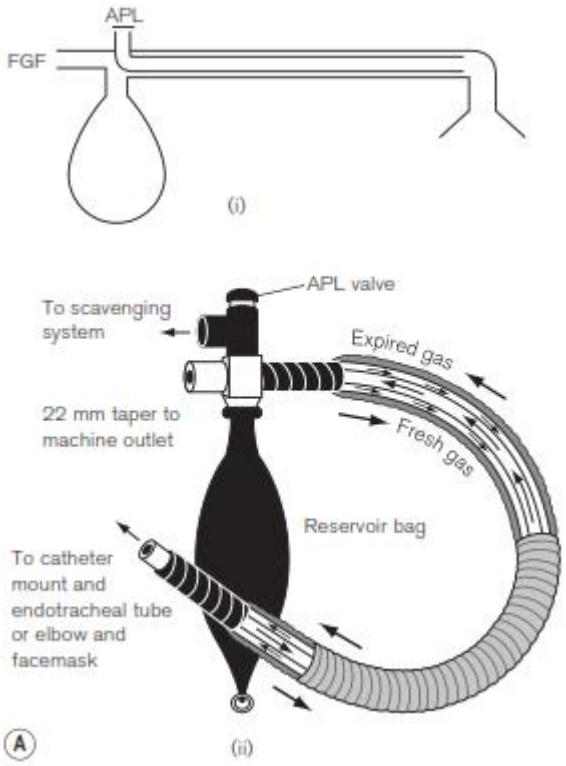
- reduces the access to the valve in head and neck surgery
- increases the drag on the mask or endotracheal tube when the valve is shrouded and connected toscavenging tubing.

The Lack system overcomes these two problems. The original version was constructed with a co-axial arrangement of breathing hoses. Exhaled gasses are passed into the oriice of the inner hose sited at the patient end of the system and then back towards the APL valve, which is now sited on the reservoir bag mount. The valve is thus conveniently sited for
Adjustment by the anaesthetist and its weight, and that of any additional scavenging attachment, is now supported by the anaesthetic machine (see Fig5.7). The system still functions as a Mapleson A system. The co-axial hosing on early models was criticised as
being too narrow and having too high a low resistance.In later models, the inner and outer breathing hose diameters were subsequently both increased, to 15 and 30 mm respectively, to overcome this problem. Another problem was that the tubing was heavy and stiff, putting a stress on the connection to the facepiece or endotracheal connection.

Lack parallel breathing system

(Fig. 5.7B)Co-axial breathing systems have particular hazards. If the inner hose were to become disconnected or to split, as has been the case, the leak may

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



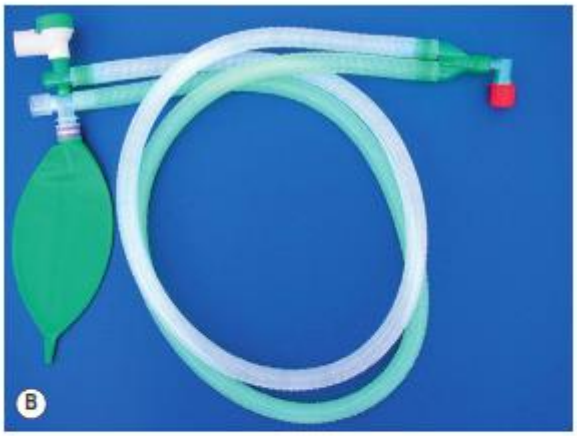


Figure 5.7 **A.** The Lack co-axial breathing system. (i) Working principles. (ii) The actual assembly. The outer corrugated hose is partly transparent so that the inner tube may be seen to be intact. The adjustable pressure-limiting valve (APL) is fitted with a shroud having a 30 mm outlet so that it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

Mapleson D system

The Mapleson D system with spontaneous respiration

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

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

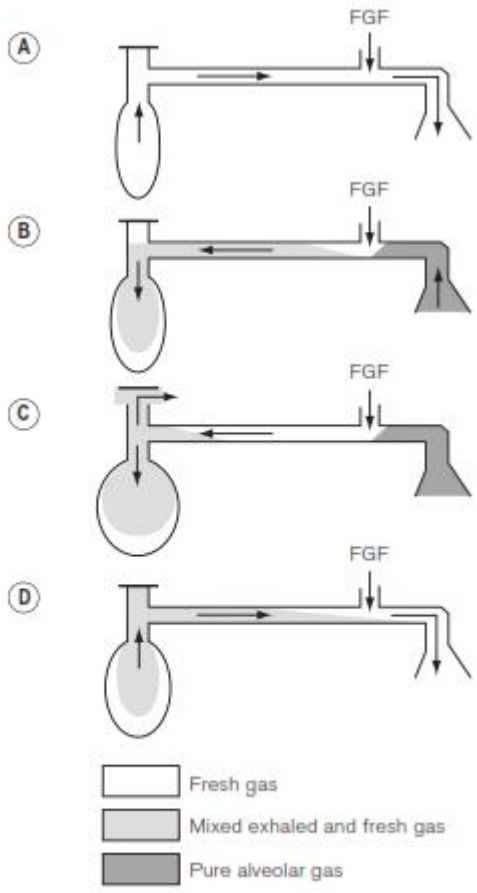


Figure 5.8 The Mapleson D system with spontaneous ventilation (see text). FGF, fresh gas flow; APL, adjustable pressure-limiting valve.

• *Expiration.* During expiration, the exhaled gasses mix with fresh gas entering the system and these pass down the wide bore hose. They initially displace any fresh gas remaining here and start to ill the reservoir bag (Fig. 5.8B). When the bag is full, the remainder of the exhaled gasses and the FGF are voided via the APL (expiratory) valve. Of the expired gasses, it is those from the patient's respiratory dead space that are voided irst, followed by alveolar gasses.

•*End-expiratory pause.* During the end-expiratory pause the fresh gas low entering the system passes down the wide bore hose, displacing some of the mixture

of exhaled gas and FGF, which is now vented out through the APL valve (Fig. 5.8C). The amount of
fresh gas occupying and thus stored in the patient end of the wide bore hose at the end of expiration, Breathing systems and their components therefore depends on the fresh gas low rate, the duration of the end-expiratory pause, and the degree of mixing (due to turbulence) of the various gaseous
interfaces within the corrugated hose.

• At the next inspiration the inhaled gasses consist initially of this stored fresh gas followed
then by the mixture of exhaled gasses and FGF that remain in the tube and possibly some of the mixture from the reservoir bag if the inhaled tidal volume is large (Fig. 5.8D). However, there are practical problems with this concept, since the expiratory pause may be short, particularly during spontaneous breathing (when volatile anaesthetics only are used with minimal opioid supplementation). In this case the fresh gas low needs to be suficiently high to lush the exhaled gasses downstream prior to the next inspiration. In fact, rebreathing of exhaled alveolar gas occurs unless the fresh gas low is at least 2 times and possibly up to 4 times the patient's minute ventilation. It is worthy of note that the Mapleson D system is functionally similar to a T-piece (Mapleson E). However, with a T-piece, the limb through which the ventilation occurs, if used without a reservoir bag, must be of such a length that the volume of gas in it when augmented by the volume of the fresh gas low being delivered during inspiration is no less than that of the patient's tidal volume, otherwise dilution of anaesthetic by entrained air will occur.

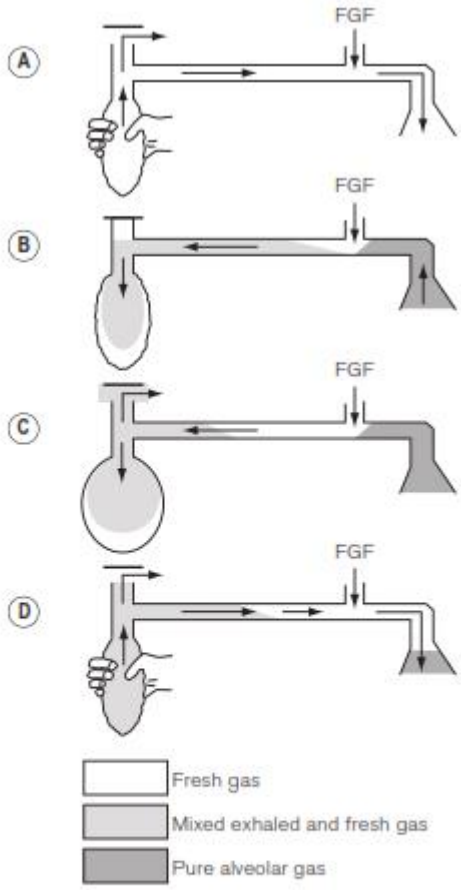


Figure 5.9 Mapleson D system with manual ventilation. **A.** The first inspiration; note that the APL valve is forced open. **B.** Early exhalation; the APL valve is closed and the partially collapsed reservoir bag is filling. **C.** Late exhalation/ expiratory pause; mixed gas is vented from the system.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

Mapleson D system with controlled or assisted ventilation

•*The first inspiration.* As the bag (full of fresh gas) is squeezed, the fresh gas low entering the system as well as fresh gasses stored in the wide-bore breathing hose pass to the patient. At the same time some gasses from the reservoir bag are

lost through the partially open APL (expiratory) valve (Fig. 5.9A).

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

. *Thus, to prevent rebreathing* in the Mapleson D system during both spontaneous and controlled ventilation, the fresh gas low must be suficiently high enough to:

- purge the breathing hose of exhaled gasses
- supplement the stored fresh gas in this breathing hose so that any mixed gas in the reservoir bag is prevented from reaching the patient. The amount of fresh gas required will always be greater than the patient's minute volume and will depend largely on the expiratory pause. The longer the pause, the more effective will be the ability of the fresh gas to purge the breathing hose of expired gas. However, in practice during controlled ventilation, *deliberate use is often made of functional rebreathing*. Theoretically,if slow ventilation rates (with long expiratory pauses) and large tidal volumes are chosen, then sufficient expiratory time will elapse to allow a modest fresh gas low to ill the proximal part of the system with suficient fresh gas to provide alveolar ventilation. This will enter the lungs first, followed by a mixture of previously expired gasses which will then occupy the patient's anatomical dead space. Hence, theoretically, it should be possible to reduce the fresh gas low to the volume required for alveolar ventilation.

In practice, there is turbulent mixing of the various gaseous interfaces so that alveolar gas is widely distributed (and diluted). Even so, provided *suficiently large controlled minute ventilation* is delivered so that most of the fresh gas low reaches the alveoli, adequate alveolar ventilation will occur with fresh gas low rates of 70% of the anticipated minute ventilation since, as mentioned above, some rebreathing is acceptable.

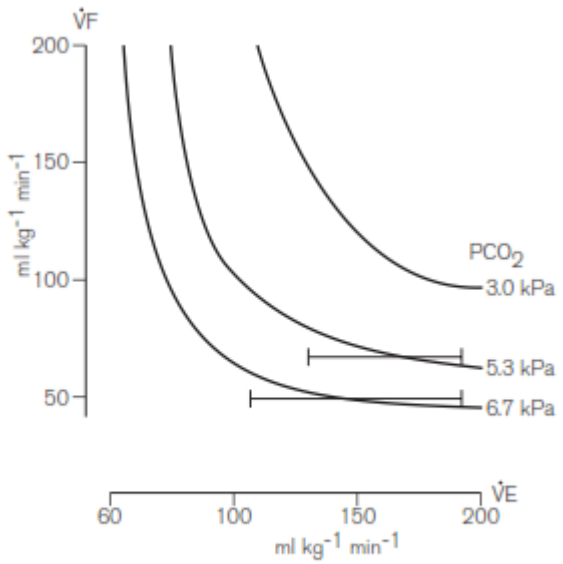


Figure 5.10 Isopleths showing the relationship between fresh gas flow (V_f), minute ventilation (V_e) and various alveolar carbon dioxide tensions.

Fig. 5.10 demonstrates this as well as the fact that the arterial carbon dioxide tension remains fairly constant for any given fresh gas low rate despite alterations in minute ventilation. Mapleson D systems are thus able to make eficient use of fresh anaesthetic gasses during controlled ventilation and could have considerable cost saving beneits.

Fig. 5.11 shows how the Mapleson D system may be employed with an automatic ventilator. The reservoir bag is removed and replaced with a standard length of corrugated hose of suficient capacity to accommodate the air or oxygen that is delivered by the ventilator, and therefore prevents it reaching the patient in place of the intended anaesthetic gasses.

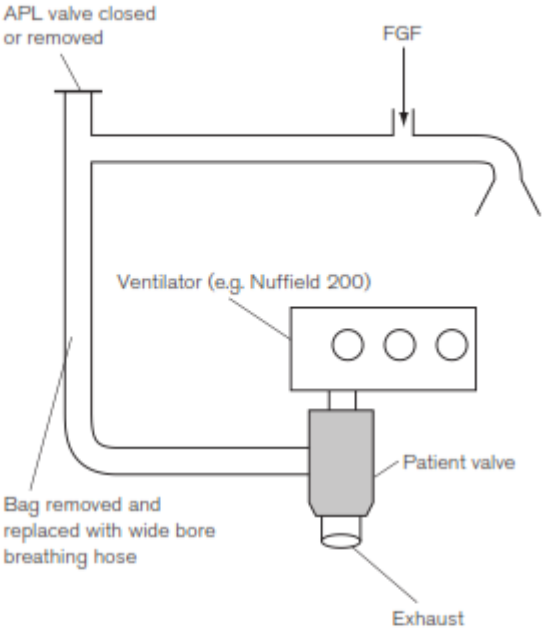


Figure 5.11 Controlled ventilation with a Mapleson D system using a ventilator. Note that with the employment of a ventilator that operates as a 'gas piston', the former must be separated from the breathing system by a suitably long piece of breathing hose (with an internal volume of at least 500 ml).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

HUMIDIFIERS

Humidity Humidity is used to describe the amount of water vapour in air or gas. The mass of water vapour in the gas is the absolute humidity (g m ?3). The maximum amount of humidity that gas can contain is limited by temperature (Fig. 11.5). At the maximum humidity for a particular temperature, the gas is said to be saturated with water vapour, and the level of humidity is the humidity at saturation. The relative humidity (RH; %) is the absolute humidity of the gas at a particular temperature as a percentage of the humidity at saturation at the same temperature.

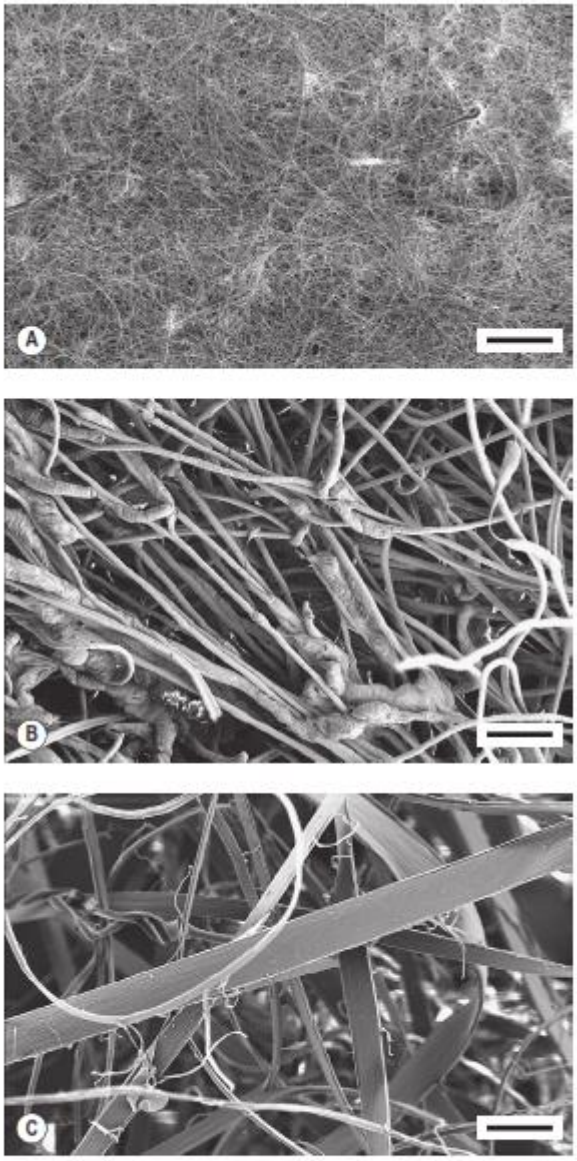


Figure 11.3 Scanning electron microscope photographs of the surface of different types of filter material. **A.** Glass fibre; **B.** tribocharged electrostatic; **C.** fibrillated electrostatic. Note the 100 µm scale marker in the bottom right corner of the photographs. Bacteria are about 1 µm in diameter.

Room air at 22°C typically has an absolute humidity level of approximately 10 g m 278 . The humidity at saturation of air At 22°C is approximately 20 gm , so that the room air has a relative humidity of approximately 50% RH. If the air is cooled, a point is reached at about 11°C when the absolute humidity level equals the humidity at saturation, and hence the relative humidity is 100%. If the air cools to an even lower temperature, condensation will occur. If room air is inspired, the air is warmed to 37°C by the upper air passages by the time it reaches the lungs, the humidity is increased from 10 to 44 g m (BTPS conditions: body temperature and pressure, saturated).

The difference between the two (34 Gm) is the humidity

deficit: humidity must be added by the airways toreduce this deicit to 0 g m
. If the room air is warmed from 22 to 37°C without any humidiication, the relative humidity will fall to 100 □ (10 □ 44) □ 23%. The massic enthalpy (latent heat) of evaporation of water is 2.4 kJ g
. To saturate inspired gasses, which have a low level of humidity, a considerable proportion of the body's heat production must be used (up to one-third for a neonate). This can then lead to a fall in the patient's core temperature ofMore than 1°C.

Humidification requirements

The level of humidity acceptable in gasses delivered to patients whose upper airways have been bypassed depends on the length of time of the bypass. For short-term use the
level of humidity may only need to be 20 g m (45% of BTPS conditions). For longer use, for example, for patients in intensive care units, the level of humidity should be at
least 33 g m (75% of BTPS conditions). One group has proposed that the level of humidity should be close to BTPS conditions (44 g m).

Humidification equipment

Gas can be humidified using either passive or active systems. Passive systems, such as heat and moisture exchangers (HMEs), rely on the patient's ability to add moisture to the inspiratory gas. Active systems, such as heated humidiiers, add water vapour to a low of gas independently of the patient. Combined passive and active devices are also now available.

- ***Passive humidification systems***

1. Heat and moisture exchangers (HMEs) HMEs return a portion of the exhaled heat and moisture to the next inspiration. If the exhaled gas is at 34°C and is saturated with water vapour (38 g m) then, even if the HME is 80% eficient, only 30 g m is returned to the patient.These devices generally consist of a transparent plastic housing so that any obstructions and secretions in the device can be seen readily. The housing contains a layer of either foam or paper that is commonly coated with a hygroscopic salt such as calcium chloride. The expired gas cools as it passes through this layer and condensation occurs, releasing the massic enthalpy of vaporization, which is partly retained by the HME layer. The hygroscopic salt absorbs water vapour, hence reducing the relative humidity of the gas to below saturation level, although some moisture is always lost into the breathing system.
During inspiration, the humidity level of the gas entering the HME is usually low, so that the condensate evaporates using the absorbed heat, which also

warms the gas. The hygroscopic salt, to which the water molecules are loosely bound, releases the water molecules when the water vapour pressure is low. The inspired gas is, therefore, warmed and humidiied to an extent that depends on the moisture content of the expired gas, and, hence, on the patient's core temperature. A layer of ilter material may also be added to the device (heat and moisture exchange ilters, HMEFs). Devices are also available that consist only of a layer of ilter material. An electrostatic ilter layer used on its own has a very low moisture-conserving performance, but a pleated hydrophobic filter layer used on its own can return some moisture as a temperature gradient builds up within the pleats, allowing condensation during expiration and evaporation during inspiration.

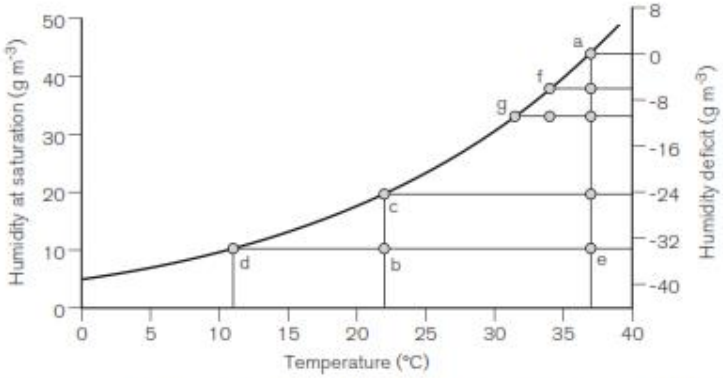


Figure 11.5 Humidity at saturation against temperature. Humidity deficit indicates the humidity that must be added by the airways to increase the humidity to (a) BTPS conditions (37°C, 44 g m^{-3}, 100% relative humidity). (b) Typical room air (22°C, 10 g m^{-3}, 50% RH); (c) room air saturated with water vapour (22°C, 20 g m^{-3}, 100% RH); (d) room air cooled so that condensation occurs (11°C, 10 g m^{-3}, 100% RH); (e) room air warmed to body temperature (37°C, 10 g m^{-3}, 23% RH); (f) expired gas (34°C, 38 g m^{-3}, 100% RH); (g) minimum moisture output for humidifiers intended for use with patients whose airways have been bypassed (33 g m^{-3}).

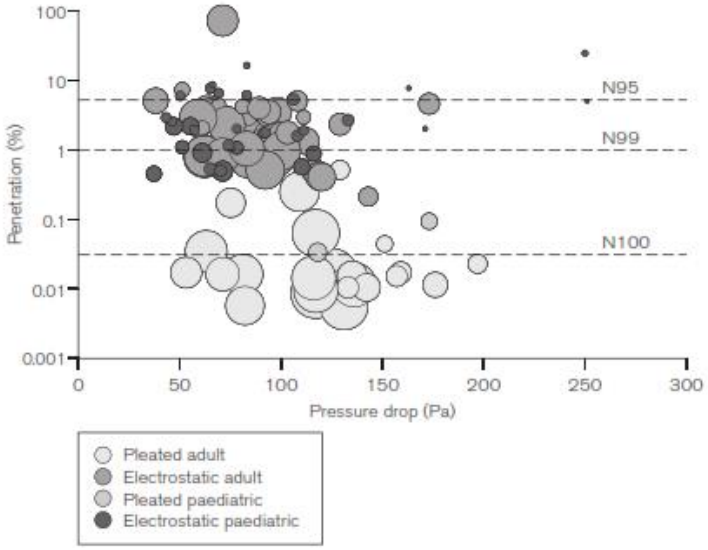


Figure 11.4 Penetration through 104 different filters when challenged with an aerosol of sodium chloride particles at the most penetrating particle size against pressure drop measured across the filter (100 Pa ≈ 1 cm H_2O). Filters were tested at flows of 15 l min^{-1} (paediatric) or 30 l min^{-1} (adult), respectively. The size of each bubble is proportional to the internal volume of the filter. An ideal filter would have low values for penetration, pressure drop and internal volume (a small 'bubble' in the lower left hand side of the figure). However, filters with low penetration (high filtration efficiency) and low-pressure drop tend to be large filters; smaller filters tend to have higher pressure drops and higher penetration values (low filtration performance). N95, N99 and N100 refer to filtration efficiencies: namely <5%, <1% and <0.03% penetration respectively.

Adapted from Wilkes AR. Heat and moisture exchangers and breathing system filters: their use in anaesthesia and intensive care. Part 1 – history, principles and efficiency. Anaesthesia *2011;**66**:31-39 with permission from Wiley-Blackwell.*

Classification of filters and heat and moisture exchangers

There are, therefore, ive broad types of ilter and/or heatand moisture exchange devices (
Fig. 11.6):

• A heat and moisture exchange-only device, without afilter layer

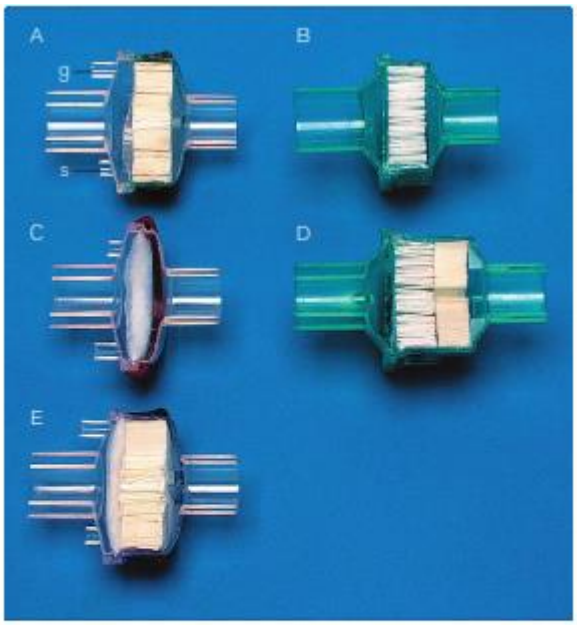


Figure 11.6 Cross-section through HMEs, HMEFs and filters. **A.** HME only; **B.** pleated hydrophobic (glass fibre) filter only; **C.** electrostatic filter only; **D.** pleated hydrophobic (glass fibre) filter and HME; **E.** electrostatic filter and HME. Note the gas sampling port (g) and the storage port (s) for the cap from the port when the port is in use.

• A filter-only device, without a heat and moisture exchange layer, which can be either: electrostatic, or pleated hydrophobic (glass ibre) type

•Combined devices with both a heat and moisture exchange layer and a ilter layer (HMEF), which can be either: electrostatic, or pleated hydrophobic (glass fibre) type.

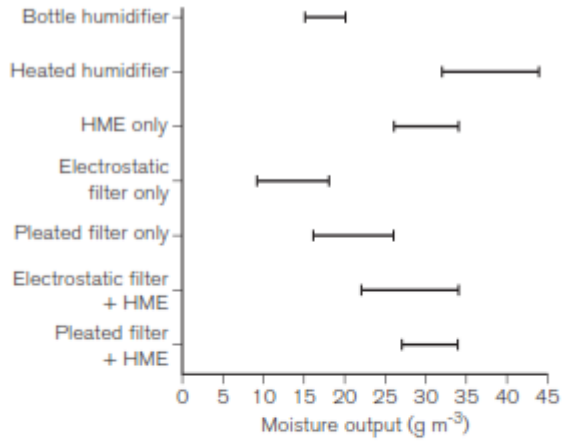


Figure 11.7 Range of humidification performance of different devices. The moisture output of bottle humidifiers depends on the ambient temperature and the degree of cooling. The moisture output of HMEs, HMEFs and filters varies for different devices and also depends on the tidal volume. In theory, a heated humidifier can be set to deliver any level of humidity if it has the appropriate controls. The values shown illustrate those that can be obtained when following the manufacturer's recommendations.

The range of typical levels of moisture output available is shown in Fig. 11.7. Devices available from some manufacturers are colour-coded to indicate particular types of device (Fig. 11.8). Filters and HMEs are available with different connectors. These include angle-pieces, catheter mounts and gas sampling ports. The gas sampling port is a female Luer-

lock connector. When the cap on the port is removed, it can be placed on a storage port (if one is available)
or, preferably a tether prevents the cap from becoming mislaid. HMEs are also available that are intended to be connected to a tracheostomy tube (Fig. 11.9).

Figure 11.8 Examples of HMEs, filters and HMEFs from one manufacturer. Note the colour-coding: blue, HME-only; yellow, filter-only; green, HMEF. Other manufacturers use different colour-coding. The different sizes of devices are intended for use with paediatric and adult patients.

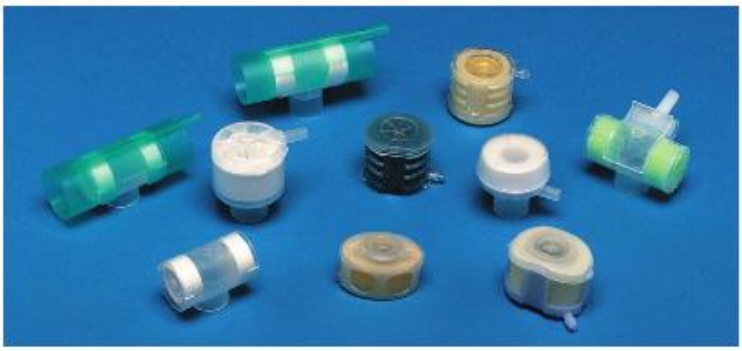


Figure 11.9 Heat and moisture exchangers intended to be attached to a tracheostomy tube. An oxygen tube can be connected to a port on some devices; some devices also have a port through which the patient's airways can be suctioned. One device has a valve which, when pressed closed, occludes, allowing the patient to speak.

2. Circle breathing systems

In these breathing systems, a portion of the exhaled gas is usually returned to the inspiratory limb of the breathing system in order to recycle the anaesthetic agents. This gas

is first passed through a container of absorbent in order to remove the exhaled carbon dioxide. This reaction generates both heat and water.
The removal of one mole of carbon dioxide from exhaled gas generates one mole of water. If all this water vaporizes, and assuming ideal gas conditions exist, an identical
volume of water vapour will replace the volume of carbon dioxide. If the exhaled gas contains 5% carbon dioxide, suficient water vapour could be produced to saturate gas
at 33°C, as saturated gas at 33°C contains 5% water vapour by volume (36 g m water vapour). This is much more than the minimum of 20 g m recommended when the
upper airways are bypassed during short-term procedure.
Hence, provided low fresh-gas lows are used with the circle breathing system (so that the humidity in the gas is not diluted excessively by the dry fresh-gas), adequate levels of humidity may be produced by the circle system alone(Fig. 11.10).

During anaesthesia, the humidity of the gas in the circle system therefore increases, although it may take up to 1 hour or more to reach a maximum level. The humidity in the breathing system will augment the moisture content Of the gas delivered To the patient From a Filter or HME sited at the patient connection port.

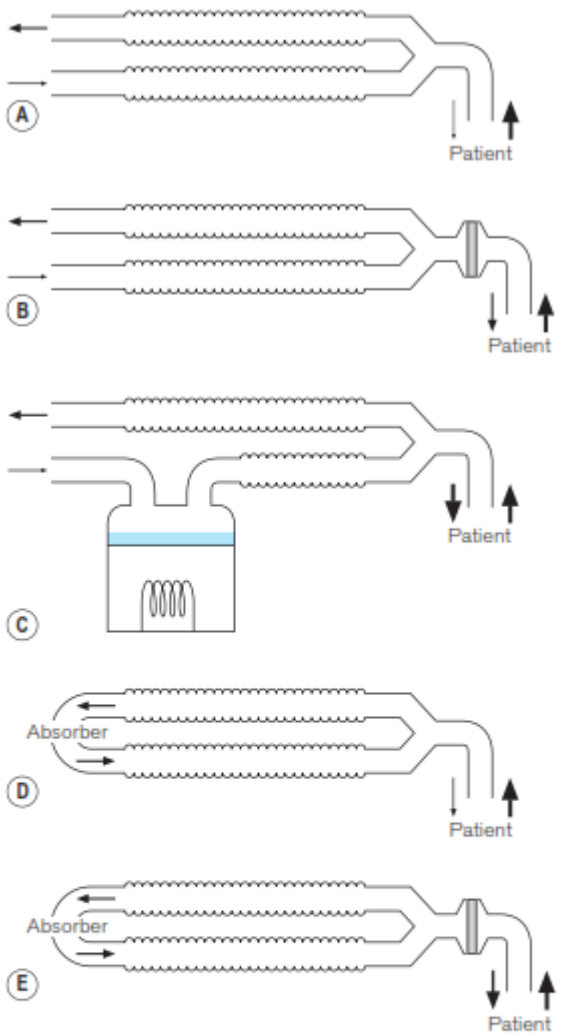


Figure 11.10 Breathing systems with different methods of humidification. **A.** Open breathing system; **B.** breathing system with HME; **C.** breathing system with heated humidifier; **D.** circle breathing system; **E.** circle breathing system with HME. The width of the arrows indicates the level of humidity at various points in the breathing system. High levels of humidity can cause condensation if an appropriate temperature is not maintained.

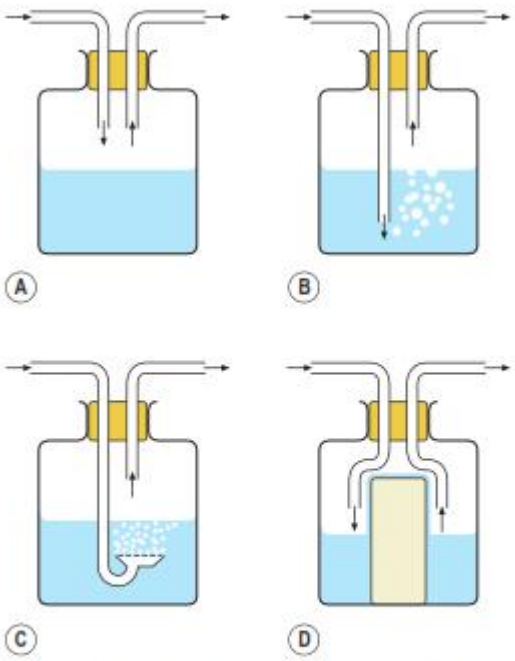


Figure 11.11 Ambient temperature or bottle humidifier. **A.** Simple bottle humidifier; **B.** bubble-through humidifier; **C.** bubble-through humidifier with sintered filter to reduce the size of bubbles and hence increase surface area for evaporation; **D.** wick humidifier.

- ***Active humidification systems***

In these devices, water vapour is added to the inspired gasses independently of the patient.

1. Bottle humidifier The simplest type of active humidifier is the bottle humidifier

(Fig.11.11). In its simplest form, gas is directed over the surface of the water (Fig. 11.11A). The water evaporates and the water vapour mixes with the gas, increasing its

humidity. The process is ineficient in that, unless the gas low is very low, there is not suficient time for the gas to become saturated with water vapour before it leaves

the humidifier. The eficiency of humidification can be improved by increasing the surface area for evaporation. This can be achieved by bubbling the gas through the water, either through a tube (Fig. 11.11B) or through a sintered filter (Fig. 11.11C), or by placing a wick into the water: water is drawn up the wick again increasing the surface area for evaporation (Fig. 11.11D). Bubbling gas through the water increases the pressure drop across the

device, and the resistance to gas low may, therefore, be unacceptably high for a spontaneously breathing patient who is required to draw gas through the humidifier.

However, in all these systems, the ambient temperature limits the maximum level of humidity that can be generated. For typical room air, the maximum level of (humidity
at saturation) that can be achieved is about 20 g m . However, because the water will cool as evaporation occurs, the level is likely to be less than this. To achieve higher levels of humidity, either the gas or the water (or both) must be heated. Alternatively, a nebulizer may be used .

Figure 11.12 The Medisize Hygrovent Gold. Water is gravity-fed into a space between a heater plate and a Goretex membrane. The water is heated and evaporates: the water vapour then passes through the Goretex membrane into the breathing system.

2. Active heat and moisture exchanger

Breathing ilters, humidiiers and nebulizers In this device, there is a heat and moisture exchange layer that retains and returns a portion of the exhaled heat and moisture as with passive versions. However, there is also a source of water that is heated within the device so that additional water vapour can be added to the inspiratory
gasses. One version of the device can be used with any HME or HMEF, and, therefore, if a filter is used, the patient
can be protected from inhaling any infective droplets (Fig. 11.12). This device adds about 5 g m to the humidity provided by the HME or filter with which it is used.

Figure 11.13 Hudson RCI Conchatherm IV heated humidifier. There are separate controls for 'TEMPERATURE' (temperature of the gas delivered to the patient) and 'TEMP GRADIENT' (difference between the temperature at the patient end of the delivery tube and the humidifier outlet).

3. Heated humidifiers

Heated humidiiers typically consist of the following (fig 11.13). Water is heated in a chamber that includes a

wick, which absorbs the water and increases the surface area for evaporation. The water vapour produced mixes with the supplied gas as it lows through the humidiier. The water lost by evaporation is either manually or automatically replenished from a reservoir. The humidiied gas then lows to the patient through a delivery tube, which contains a sensor that monitors the temperature of the gas at the
patient end. The desired temperature of the delivered humidified gas is set using a control knob on the humidiier.
If this gas is allowed to cool as it lows through the delivery tube, condensation may occur. To reduce this cooling, a heater wire in the delivery tube maintains the
temperature of the gas. The temperature of the gas at the outlet of the humidiication chamber is also monitored. The difference in temperature between the humidiication chamber and the patient connection port is set on a second control knob. If the heater wire increases the temperature of the gas as it lows through the delivery tube, the risk of
condensation forming is reduced, but the relative humidity of the gas also decreases. Alternatively, if the temperature of the gas falls, because of the settings on the two knobs,
some condensation will occur, but the gas remains fully saturated with water vapour (
Fig. 11.14). Unless a second heater wire is used, condensation will also occur in the expiratory limb. Alternatively, a water trap can be itted to collect any condensation that forms. In one type of humidiier, the user has only to select whether the patient is intubated or receiving humidified gasses via a facemask. The temperature and humidity of the gasses are then controlled by the humidiier to provide the optimum level for the patient. The humidifier controls the humidity and temperature and also takes into account the low of gas (Fig. 11.15).