Chapter 1

Equipment and monitoring

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The optimum conditions for cardiothoracic surgery have traditionally been regarded as a "still and bloodless" surgical field. Cardiopulmonary bypass (CPB) provides this by incorporating a pump to substitute for the function of the heart and a gas exchange device, the "oxygenator," to act as an artificial lung. Cardiopulmonary bypass thus allows the patient's heart and lungs to be temporarily devoid of circulation, and respiratory and cardiac activity suspended, so that intricate cardiac, vascular or thoracic surgery can be performed in a safe and controlled environment.

History

In its most basic form, the CPB machine and circuit comprises of plastic tubing, a reservoir, an oxygenator and a pump. Venous blood is drained by gravity into the reservoir via a cannula placed in the right atrium or a large vein, pumped through the oxygenator and returned into the patient's arterial system via a cannula in the aorta or other large artery. Transit through the oxygenator reduces the partial pressure of carbon dioxide in the blood and raises oxygen content. A typical CPB circuit is shown in Figure 1.1.

Cardiac surgery has widely been regarded as one of the most important medical advances of the twentieth century. The concept of a CPB machine arose from the technique of "cross-circulation" in which the arterial and venous circulations of mother and child were connected by tubing in series. The mother's heart and lungs maintained the circulatory and respiratory functions of both, whilst surgeons operated on the child's heart (Dr Walton Lillehei, Minnesota, 1953, see Figure 1.2a). Modern CPB machines (see Figure 1.2b) have evolved to incorporate monitoring and safety features in their design.

John Gibbon (Philadelphia, 1953) is credited with developing the first mechanical CPB system, which he used when repairing an atrial secundum defect (ASD). Initially, the technology was complex and unreliable and was therefore slow to develop. The equipment used in a typical extracorporeal circuit has advanced rapidly since this time and although circuits vary considerably among surgeons and hospitals, the basic concepts are essentially common to all CPB circuits.

This chapter describes the standard equipment and monitoring components of the CPB machine and extracorporeal circuit as well as additional equipment such as the suckers used to scavenge blood from the operative field, cardioplegia delivery systems and hemofilters (see Tables 1.1 and 1.2).

Tubing

The tubing in the CPB circuit interconnects all of the main components of the circuit. A variety of materials may be used for the manufacture of the tubing; these include polyvinyl chloride

Cardiopulmonary Bypass, ed. S. Ghosh, F. Falter and D. J. Cook. Published by Cambridge University Press.

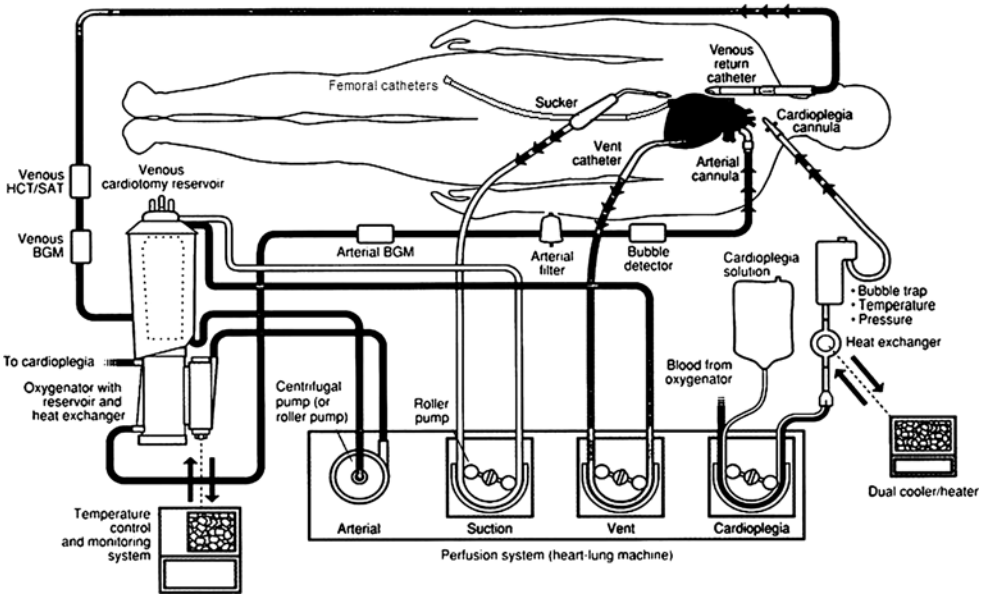

Figure 1.1. Typical configuration of a basic cardiopulmonary bypass circuit. BGM = blood gas monitor; SAT = oxygen saturation.

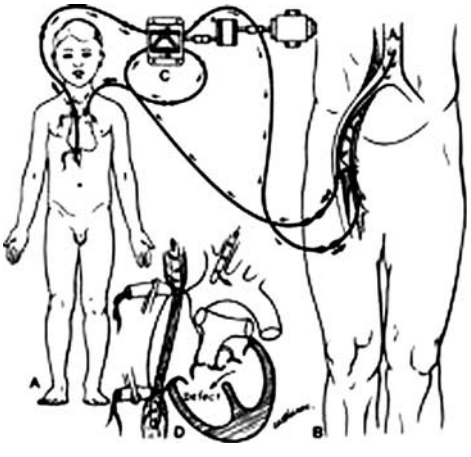

Figure 1.2a. Depiction of the method of direct vision intracardiac surgery utilizing extracorporeal circulation by means of controlled cross circulation. The patient (A), showing sites of arterial and venous cannulations. The donor (B), showing sites of arterial and venous (superficial femoral and great saphenous) cannulations. The Sigma motor pump (C) controlling precisely the reciprocal exchange of blood between the patient and donor. Close-up of the patient's heart (D), showing the vena caval catheter positioned to draw venous blood from both the superior and inferior venae cavae during the cardiac bypass interval. The arterial blood from the donor circulated to the patient's body through the catheter that was inserted into the left subclavian artery. (Reproduced with kind permission from Lillehei CW, Cohen M, Warden HE, *et al.* The results of direct vision closure of ventricular septal defects in eight patients by means of controlled cross circulation. Surg Gynecol Obstet 1955; 101: 446. Copyright American College of Surgeons.)

(PVC, by far the most commonly used), silicone (reserved for the arterial pump boot) and latex rubber. The size of tubing used at different points in the circuit is determined by the pressure and rate of blood flow that will be required through that region of the circuit, or through a particular component of the circuit (see Table 1.3).

PVC is made up of polymer chains with polar carbon-chloride (C-Cl) bonds. These bonds result in considerable intermolecular attraction between the polymer chains, making PVC a fairly strong material. The feature of PVC that accounts for its widespread use is its versatility. On its own, PVC is a fairly rigid plastic, but plasticizers can be added to make it highly flexible. Plasticizers are molecules that incorporate between the polymer chains allowing them

Figure 1.2b. Cardiopulmonary bypass machine (reproduced with kind permission of Sorin Group).

to slide over one another more easily, thus increasing the flexibility of the PVC. However, one disadvantage is that PVC tubing stiffens during hypothermic CPB and tends to induce spallation; that is, the release of plastic microparticles from the inner wall of tubing as a result of pump compressions.

Other materials used to manufacture perfusion tubing include latex rubber and silicone rubber. Latex rubber generates more hemolysis than PVC, whereas silicone rubber is known to produce less hemolysis when the pump is completely occluded, but can release more particles than PVC. As a result of this, and because of PVC's durability and accepted hemolysis rates, PVC is the most widely used tubing material. The arterial roller pump boot is the main exception to this, as the tubing at this site is constantly compressed by the rollers themselves, leading to the use of silicone tubing for this purpose.

Arterial cannulae

The arterial cannula is used to connect the "arterial limb" of the CPB circuit to the patient and so deliver oxygenated blood from the heart-lung machine directly into the patient's arterial system. The required size is determined by the size of the vessel that is being cannulated,

Table 1.1. Components of the CPB machine and the extracorporeal circuit

Equipment	Function
Oxygenator system, venous reservoir, oxygenator, heat exchanger	Oxygenate, remove carbon dioxide and cool/re-warm blood
Gas line and FiO_2 blender	Delivers fresh gas to the oxygenator in a controlled mixture
Arterial pump	Pumps blood at a set flow rate to the patient
Cardiotomy suckers and vents	Scavenges blood from the operative field and vents the heart
Arterial line filter	Removes microaggregates and particulate matter >40 μm
Cardioplegia systems	Deliver high-dose potassium solutions to arrest the heart and preserve the myocardium
Cannulae	Connect the patient to the extracorporeal circuit

Table 1.2. Monitoring components of the CPB machine and the extracorporeal circuit

Monitoring device	Function
Low-level alarm	Alarms when level in the reservoir reaches minimum running volume
Pressure monitoring (line pressure, blood cardioplegia pressure and vent pressure)	Alarms when line pressure exceeds set limits
Bubble detector (arterial line and blood cardioplegia)	Alarms when bubbles are sensed
Oxygen sensor	Alarms when oxygen supply to the oxygenator fails
S_aO_2, S_vO_2, and hemoglobin monitor	Continuously measures these levels from the extracorporeal circuit
In-line blood gas monitoring	Continuously measures arterial and venous gases from the extracorporeal circuit
Perfusionist	Constantly monitors the cardiopulmonary bypass machine and the extracorporeal circuit

Table 1.3. Tubing sizes commonly used in different parts of the extracorporeal circuit (adults only)

Tubing size	Function
3/16″ (4.5 mm)	Cardioplegia section of the blood cardioplegia delivery system
1/4″ (6.0 mm)	Suction tubing, blood section of the blood cardioplegia delivery system
3/8″ (9.0 mm)	Arterial pump line for flow rates <6.7 l/minute, majority of the arterial tubing in the extracorporeal circuit
1/2″ (12.0 mm)	Venous line, larger tubing is required to gravity drain blood from the patient

as well as the blood flow required. The ascending aorta is the most common site of arterial cannulation for routine cardiovascular surgery. This is because the ascending aorta is readily accessible for cannulation when a median sternotomy approach is used and has the lowest associated incidence of aortic dissection (0.01–0.09%). After sternotomy and exposure, the surgeon is able to assess the size of the aorta before choosing the most appropriately sized cannula (see Table 1.4).

Table 1.4. Arterial cannulae flow rates in relation to type/size

	Size		
Cannulae	French gauge	mm	**Flow rate (l/minute)**
DLP angled tip	20	6.7	6.5
	22	7.3	8.0
	24	8.0	9.0
DLD straight tip	21	7.0	5.0
	24	8.0	6.0
Sarns high flow angled tip	15.6	5.2	3.5
	19.5	6.5	5.25
	24	8.0	8.0
Sarns straight tip	20	6.7	5.9
	22	7.3	6.0
	24	8.0	6.0

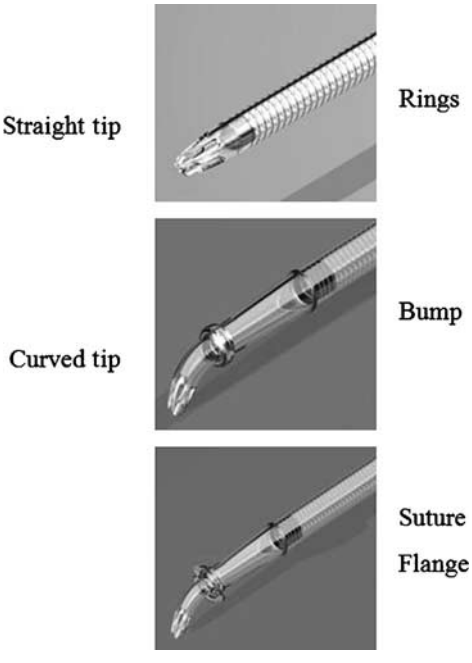

Figure 1.3. Commonly used arterial cannulae. (Reproduced with kind permission from Edwards Lifesciences.)

Thin-walled cannulae are preferred, as they present lower resistance to flow because of their larger effective internal diameter. This leads to a reduction in arterial line pressure within the extracorporeal circuit and increased blood flow to the patient.

Arterial cannulae with an angled tip are available. These direct blood flow towards the aortic arch rather than towards the wall of the aorta; this may minimize damage to the vessel wall. In addition, cannulae with a flange near the tip to aid secure fixation to the vessel wall and cannulae that incorporate a spirally wound wire within their wall to prevent "kinking" and obstruction are commonly used (see Figure 1.3).

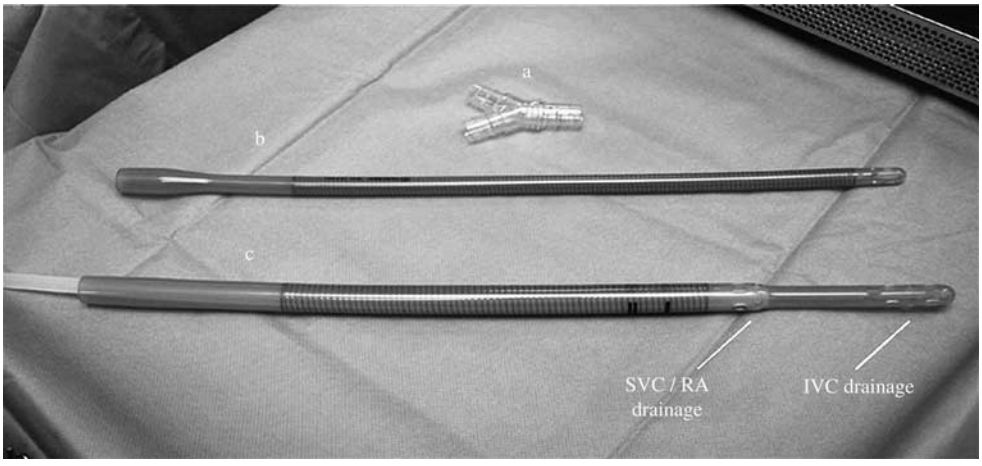

Figure 1.4. Commonly used venous cannulae: (a) Y-connector to connect single-stage cannulae; (b) single-stage cannula; (c) two-stage cannula. RA, right atrial; SVC, superior vena cava; IVC, inferior vena cava.

Venous cannulae

Venous cannulation for CPB allows deoxygenated blood to be drained from the patient into the extracorporeal circuit. The type of venous cannulation used is dependent upon the operation being undertaken. For cardiac surgery that does not involve opening the chambers of the heart, for example, coronary artery bypass grafts (CABG), a two-stage venous cannula is often used. The distal portion, i.e., the tip of the cannula, sits in the inferior vena cava (IVC) and drains blood from the IVC through holes around the tip. A second series of holes in the cannula, a few centimeters above the tip, is sited in the right atrium, to drain venous blood entering the atrium via the superior vena cava (SVC).

An alternative method of venous cannulation for CPB is bicaval cannulation – this uses two single-stage cannulae that sit in the inferior and superior vena cavae, respectively. The two single-stage cannulae are connected using a Y-connector to the venous line of the CPB circuit. Bicaval cannulation is generally used for procedures that require the cardiac chambers to be opened, as the two separate pipes in the IVC and SVC permit unobstructed venous drainage during surgical manipulation of the dissected heart and keep the heart completely empty of blood (see Figure 1.4).

The femoral veins may also be used as a cannulation site for more complex surgery. In this instance, a long cannula, which is in essence an elongated single-stage cannula, may be passed up the femoral vein into the vena cava in order to achieve venous drainage.

As with arterial cannulation, the size of the cannulae will depend on the vessels being cannulated as well as the desired blood flow. It is important to use appropriately sized cannulae in order to obtain maximum venous drainage from the patient so that full flow can be achieved when CPB is commenced.

Pump heads

There are two types of pumps used in extracorporeal circuits:

1. Those that produce a flow – roller pumps.
2. Those that produce a pressure – centrifugal pumps.

Figure 1.5. (a) Line drawing of a roller pump; (b) a roller pump. (Reproduced with kind permission from Sorin Group.)

Roller pumps

Initial technology developed in the mid twentieth century used non-pulsatile roller pumps in CPB machines. This technology has not changed greatly over the past 50 years.

Roller pumps positively displace blood through the tubing using a peristaltic motion. Two rollers, opposite each other, "roll" the blood through the tubing. When the tubing is

intermittently occluded, positive and negative pressures are generated on either side of the point of occlusion. Forward or retrograde flow of blood can be achieved by altering the direction of pump head rotation; thus roller pumps are commonly used as the primary arterial flow pump as well as for suction of blood from the heart and mediastinal cavity during CPB to salvage blood. Roller pumps are relatively independent of circuit resistance and hydrostatic pressure; output depends on the number of rotations of the pump head and the internal diameter of the tubing used (see Figure 1.5a,b).

This type of positive displacement pump can be set to provide pulsatile or non-pulsatile (laminar) flow. Debate over the advantages and disadvantages of non-pulsatile or pulsatile perfusion during cardiopulmonary bypass still continues. Non-pulsatile perfusion is known to have a detrimental effect on cell metabolism and organ function. The main argument in favor of pulsatile perfusion is that it more closely resembles the pattern of blood flow generated by the cardiac cycle and should therefore more closely emulate the flow characteristics of the physiological circulation, particularly enhancing flow through smaller capillary networks in comparison to non-pulsatile perfusion. The increased shear stress from the changing positive and negative pressures generated to aid pulsatile perfusion may, however, lead to increased hemolysis. Roller pumps have one further disadvantage: sudden occlusion of the inflow to the pump, as a result of low circulating volume or venous cannula obstruction, can result in "cavitation," the formation and collapse of gas bubbles due to the creation of pockets of low pressure by precipitous change in mechanical forces.

Centrifugal pumps

In 1973, the Biomedicus model 600 became the first disposable centrifugal pump head for clinical use. The Biomedicus head contains a cone with a metal bearing encased in an outer housing, forming a sealed unit through which blood can flow. When in use the head is seated on a pump drive unit. The cone spins as a result of the magnetic force that is generated when the pump is activated. The spinning cone creates a negative pressure that sucks blood into the inlet, creating a vortex. Centrifugal force imparts kinetic energy on the blood as the pump spins at 2000–4000 rpm (this speed is set by the user). The energy created in the cone creates pressure and blood is then forced out of the outlet. The resulting blood flow will depend on the pressure gradient and the resistance at the outlet of the pump (a combination of the CPB circuit and the systemic vascular resistance of the patient). Flow meters are included in all centrifugal pumps and rely on ultrasonic or electromagnetic principles to determine blood flow velocity accurately (see Figure 1.6a–c).

Despite extensive research, there is little evidence to show any benefit of one type of pump over another in clinical practice. Centrifugal pumps may produce less hemolysis and platelet activation than roller pumps, but this does not correlate with any difference in clinical outcome, including neurological function. They are certainly more expensive (as the pump head is single use) and may be prone to heat generation and clot formation on the rotating surfaces in contact with blood. In general, they are reserved for more complex surgery of prolonged duration, during which the damage to blood components associated with roller pumps may be theoretically disadvantageous.

Reservoirs

Cardiotomy reservoirs may be hardshell or collapsible. Hardshell reservoirs are most commonly used in adult cardiac surgery; collapsible reservoirs are still used by some institutions

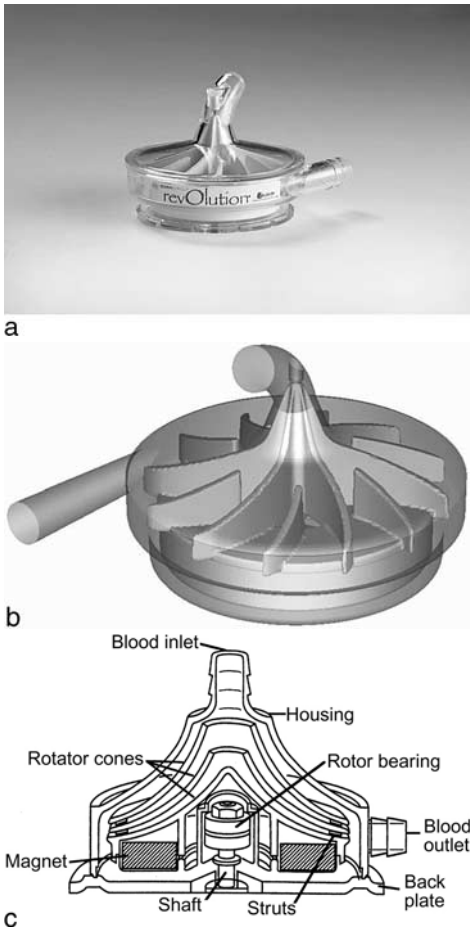

Figure 1.6. (a) Centrifugal pump. (b) Schematic diagram of centrifugal pump. (c) Schematic cut through centrifugal pump. (a, b Reproduced with kind permission from Sorin Group.)

for pediatric and adult cases. Hardshell reservoirs usually comprise of a polycarbonate housing, a polyester depth filter and a polyurethane de-foamer. The reservoir component of the CPB circuit therefore provides high-efficiency filtration, de-foaming and the removal of foreign particles (see Figure 1.7).

The reservoir acts as a chamber for the venous blood to drain into before it is pumped into the oxygenator and permits ready access for the addition of fluids and drugs. A level of fluid is maintained in the reservoir for the duration of CPB. This reduces the risks of perfusion accidents, such as pumping large volumes of air into the arterial circulation if the venous return to the CPB machine from the patient is occluded for any reason.

Blood that is scavenged from the operative field via the suckers is returned to the reservoir. The salvaged blood is mixed with air and may contain tissue debris. It is therefore vital for this blood to be filtered through the reservoir before being pumped to the patient. The reservoir is constantly vented to prevent the pressure build-up that could occur if the suckers were left running at a high level for the duration of the procedure. The salvaged blood from the vents that the surgeon uses to prevent the heart from distending during CPB also returns to the reservoir.

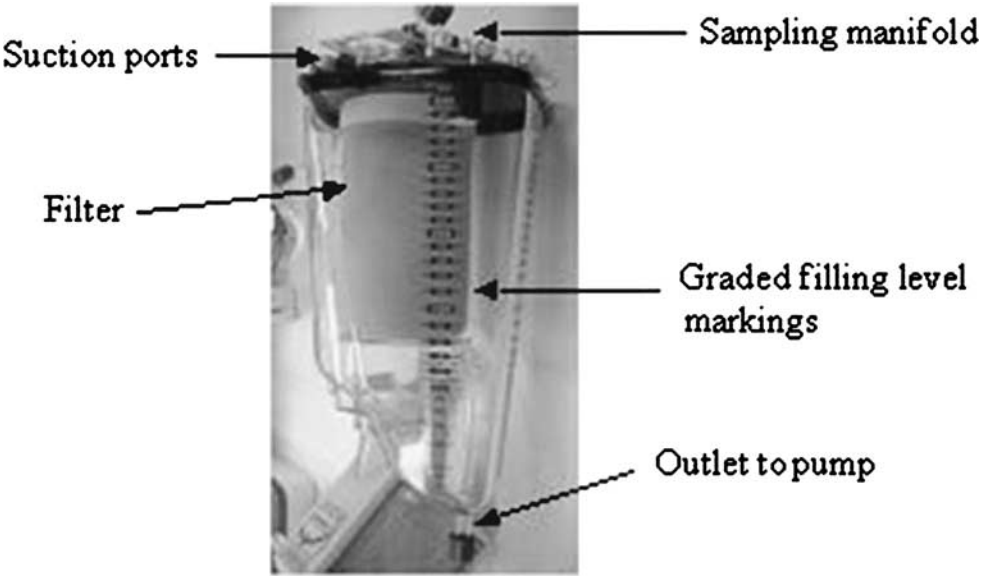

Figure 1.7. Reservoir in CPB circuit.

Oxygenators

The present success of cardiac surgery relies heavily on extracorporeal perfusion techniques employing an efficient gas exchange mechanism: the oxygenator. The requirements of the oxygenator include efficient oxygenation of desaturated hemoglobin and simultaneous removal of carbon dioxide from the blood. The oxygenator therefore acts as an artificial alveolar-pulmonary capillary system.

Gas exchange is based on Fick's Law of Diffusion:

$$\text{Volume of Gas diffused} = \frac{\text{Diffusion coefficient} \times \text{Partial pressure difference}}{\text{Distance to travel}}$$

The oxygenator provides an interface of high surface area between blood on one side and gas on the other. The distance gas has to travel across the interface is minimized by constructing the membrane from very thin material.

In the early 1950s, attempts were made to oxygenate the blood using techniques such as cross circulation between related humans, or using animal lungs for patients undergoing open heart surgery. In 1955, DeWall and Lillehei devised the first helical reservoir to be used; this was an early form of the bubble oxygenator. One year later, in 1956, the rotating disc oxygenator was developed. In 1966, DeWall introduced the hardshell bubble oxygenator with integral heat exchanger. Subsequently, Lillehei and Lande developed a commercially manufactured, disposable, compact membrane oxygenator.

Currently, most commonly used oxygenators are membrane oxygenators with a microporous polypropylene hollow fiber structure. The membrane is initially porous, but proteins in blood rapidly coat it, preventing direct blood/gas contact. The surface tension of the blood also prevents plasma water from entering the gas phase of the micropores during CPB and prevents gas leakage into the blood phase, thus reducing microemboli. However, after several hours of use, evaporation and condensation of serum leaking through micropores leads to