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The Membrane Lung

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exitus acta probat

INTRODUCTION

The first clinical use of a membrane lung was reported in 1958 by Clowes[2]. Until then, extracorporeal gas exchange had been accomplished by devices using large direct blood–gas contact surfaces, such as bubble and filming.

The initial impetus to membrane lung development was an intuitive belief that since the human lung effects gas exchange by diffusion through the alveolar membrane, which avoids direct blood–gas contact, nature's way must be physiologically preferable and should be copied.

More objective evidence favouring membrane lungs, indicating reduced trauma to the blood and the patient (haemolysis, denaturation of plasma proteins, 'pump lung'), gradually became available[11,12].

But the strongest stimulus was the more recent discovery that clinically effective, closed-chest, partial perfusion with membrane oxygenation could be performed safely for days, rather than hours, and could be used successfully to treat patients in acute pulmonary insufficiency who were otherwise unsalvageable[6,7].

It has also been clinically demonstrated[1] that the same technique is capable of reversing the vicious circle of cardiogenic shock and

gives reasonable hope of significantly lowering the present mortality rate of around 90 per cent.

The current energetic and widespread research and development efforts in the field of membrane perfusion systems are a natural and welcome consequence of those experiences.

There are presently four designs of membrane lungs in routine clinical use. They are the General Electric–Peirce[13,14], the Landé–Edwards[10], the Kolobow[9] and the Bramson membrane lung[5].

Because of our extensive clinical experience with the latter, and also because it was, according to Drinker[3], the first practical membrane system in routine use in surgery since 1965, it forms the subject of the detailed description of design and function presented in this chapter. It has been used for over 500 open-heart operations and for treating 18 cases of acute pulmonary insufficiency by prolonged partial perfusions lasting from less than 1 day to more than 12 days (*see* Chapter 18).

ESSENTIAL AND/OR DESIRABLE CHARACTERISTICS OF SYSTEMS FOR THE TOTAL OR PARTIAL PERFUSION OF ADULTS

Haemodynamic characteristics

For total perfusion of adults the flow rate obtainable (i.e. the maximum venous drainage by gravity) is usually 3–4.5 litres/min. However, a young patient weighing 90 kg may produce a venous outflow up to 6 litres/min. The system should therefore be capable of handling such a flow rate. The pressure gradient through the artificial lung should not (for practical reasons and to avoid excessive haemolysis) exceed 150 mmHg at maximum flow.

To permit meaningful performance comparisons between different artificial lungs, Galletti[4] has defined 'rated flow' as the perfusion flow at which an artificial lung is capable of raising the oxygen saturation of venous blood with an haematocrit of 40 per cent from 65 per cent to an arterial oxygen saturation of 95 per cent. 'Rated oxygen transfer' is that which occurs at rated blood flow, and at 65 per cent venous oxygen saturation. For total perfusion of adults, the rated flow needed has been found in practice to be at least 3.5 litres/min.

Gas exchange

The oxygen transfer capacity should be between 200 and 300 ml/min. At 6 litres/min blood flow this represents a 20–30 per cent step-up in

oxygen saturation (assuming the haemoglobin is 12.4 g per cent). This transfer capacity varies with the blood flow rate and, inversely, with the venous oxygen saturation.

The carbon dioxide excretion rate, when using 100 per cent oxygen in the gas circuit, should be approximately equal to the rate of oxygen uptake.

Priming volume

The priming volume of the complete extracorporeal circuit should of course be as small as possible. Usually, 1500 ml is an attainable goal (this does not include any reserve held available in the priming reservoir to compensate for surgical or other fluid losses).

Minimum haemolysis and other blood damage

The amount of plasma-free haemoglobin (PFH) formed per hour depends upon the haemodynamics of the system. It cannot be measured when a patient is in circuit, since the amount eliminated by his kidneys is unknown. However, the level of PFH should not exceed 60–80 mg per cent during perfusion. Platelet loss and plasma protein denaturation depend upon the blood contact surfaces and on the streamline geometry.

Gas and particulate emboli

Complete and rapid elimination of gas bubbles from the extracorporeal circuit must be feasible during priming, and efficient bubble traps in the circuit must be capable of handling any bubbles entering accidentally during perfusion — for example, through a leaky venous cannulation. Any foam from coronary return suction must be eliminated by incorporating a defoaming system in the coronary suction return line.

Filters capable of removing particulate emboli should be inserted both in the coronary suction line and the arterial line to the patient. (Such emboli may consist of dust, fat, clumped platelets or debris from the surgical field).

Constant extracorporeal volume

For ease of perfusion management, it is important (especially with membrane lungs in which the volume is not defined by a visible

blood level) that the extracorporeal blood volume be essentially constant. Otherwise, when significant changes occur in the patient's arterial or venous pressure, the perfusionist will be in doubt whether volume changes in the artificial lung are responsible for the loss or gain in the patient's blood volume, or whether there are other causes (vasodilation or surgical losses, etc.). His correct reaction depends on that knowledge. Constant volume also simplifies automatic arterial pump control.

Integral heat exchange

Since the patient's temperature must be controllable by a heat exchange device, it is clearly preferable if this is achieved without additional equipment and/or blood contact surfaces. Integral heat exchange within the lung is therefore desirable.

Recirculation

Some artificial lungs cannot achieve adequate gas exchange by a single passage of blood through them. In such cases recirculation is adopted, which means that the flow rate through the lung is greater than the perfusion flow to the patient. There are, however, several disadvantages to such an arrangement. It increases the haemodynamic pressure drop through the lung, it usually necessitates more than one pump in the circuit, it increases the haemolysis and it complicates the automatic pump control.

Automatic pump control

The highest rate of venous flow obtainable by gravity from the patient determines the maximum perfusion rate. It is nearly always desirable to use that rate. But since it varies with the patient's condition and many other factors, close and constant attention to the 'manual' pump speed control is necessary. This can become an almost intolerable burden on the perfusionist, especially when the procedure lasts several days. By automating this control so that the pumping rate is always equal to the venous outflow rate, perfusion management is simplified, not only because continuous attention to the pump control is unnecessary, but also because minor or major changes in venous flow rate become instantly obvious by the automatic change in pump speed. Remedial action, if required, can then be taken without delay.

Disposability

It is clearly desirable that hospitals should save the skilled man-hours required after each use to disassemble and reassemble, test and sterilize a non-disposable artificial lung. Hence, subject to adequate performance, a disposable 'cartridge' which comes in a sterile package is a keenly desired solution which has already been attained by several lung developers[9,10,14].

Minimal thrombogenicity of blood contact surfaces

This quality is of great importance when membrane lungs are used for prolonged perfusions because it permits the use of low concentrations of heparin, which in turn reduces the danger of uncontrollable bleeding.

MEMBRANE LUNG SYSTEMS COMPARED WITH DIRECT BLOOD–GAS CONTACT SYSTEMS

Advantages

(1) The most important advantage is the low overall time-related trauma to the patient demonstrated by the feasibility of prolonged life support (up to 19 days). This can be regarded as indirect evidence of significantly lower morbidity, even after short perfusions.
(2) Specifically, the *in vitro* haemolysis rate (8.2 mg per cent/h) of the Bramson membrane lung was respectively 9.7 per cent, 8.4 per cent and 14.7 per cent of the rates measured in strictly comparable 6-hour tests with three conventional oxygenators (one filming and two bubbling)[15].
(3) A constant blood volume is attainable in an extracorporeal circuit which comprises a membrane lung. Good perfusion management is thereby facilitated.
(4) Constant blood volume also simplifies automatic pump control, which in turn insures utilization at all times of the maximum available venous flow rate from the patient.
(5) By the same token, changes in pump speed give instant and obvious warning of physiological or technical irregularities which may call for remedial action.
(6) Membrane lungs can be designed with integral heat exchange,

thus avoiding additional blood–contact surfaces and/or equipment for temperature control.

(7) Foam cannot form within the membrane lung (so long as blood pressure is not allowed to fall below atmospheric).

Disadvantages

(1) If not totally disposable, the processing and testing of a membrane lung after each use is time consuming and requires skilled personnel.

(2) If disposable, membrane lungs currently cost considerably more than disposable non-membrane lungs of comparable capacity.

RATIONALE FOR THE USE OF A MEMBRANE LUNG SYSTEM

Now, after about 15 years of research and development, the rationale for the use of membrane lungs may be summarized.

For prolonged perfusions membrane lungs must be used because so far no other type of artificial lung has been clinically successful when circulatory support has been needed for several days.

For cardiac surgery—in which perfusions last 3 to 4 hours at the most—bubbling and filming oxygenators are being used routinely and successfully. Nevertheless, a reasonable inference of time-related blood damage (which after short perfusions does not prevent, but may delay, full recovery) can be drawn from the fact that patients have never survived non-membrane perfusions lasting days, rather than hours.

The question whether to use membrane lungs for cardiac surgery, then, becomes a matter of balancing this indirect evidence against such mundane factors as cost and convenience. If and when these latter factors become favourable to the membrane lung, a rationale for its universal use will no doubt be considered established.

PHYSICS OF GAS EXCHANGE THROUGH MEMBRANES AND LIQUIDS

When a liquid flows along a solid surface, a semi-stagnant boundary layer is formed in which the flow velocity varies from zero at the surface to that of the general flow at a small distance from the surface. That distance is the thickness of the 'boundary layer'.

When gas diffuses through a membrane into or out of blood it must also diffuse through the boundary layer, whose resistance is sometimes greater than that of the membrane.

Fick's law states, in effect, that the volume of gas diffusing through a layer of plasma in unit time is proportional to (1) the solubility of the gas in plasma; (2) the gas partial pressure difference (gradient) between the two sides of the layer; and (3) the area of the layer; and it is inversely proportional to the thickness of the layer. The layer must, therefore, be kept as thin as possible, which in practice can be achieved by providing gently turbulent flow, or 'mixing', through thin blood passages.

Gas diffusion through a membrane obeys the same law.

THE BRAMSON MEMBRANE HEART–LUNG MACHINE

Figure 3.1 shows the system in actual clinical use for prolonged circulatory support.

Membranes

The membranes in the lung are made of silicone rubber* impregnated into an open-weave fibreglass, supporting fabric, .002 inch thick, with an overall thickness of .005 to .0055 inch.

Gas-to-gas diffusion tests of these membranes have indicated an oxygen transmission of approximately 150 ml per square metre per minute per atmosphere and carbon dioxide transmission of 780 ml per square metre per minute per atmosphere.

The partial pressure gradients available in an artificial lung to which 100 per cent O_2 is supplied (in excess) are about 650 mmHg for oxygen and about 50 mmHg for CO_2 (since the P_{CO_2} of the blood should not exceed about 50 mmHg and cannot of course be less than zero on the gas side of the membrane).

This apparent 13 to 1 (partial pressure gradient) factor operating against the CO_2 transfer as compared with the O_2 transfer is eliminated in practice by two fortuitous facts: (1) Since CO_2 is about 30 times more soluble in plasma than O_2, the resistance of the boundary layer to CO_2 diffusion is negligible; (2) The diffusion coefficient of the membrane for CO_2 is over 5 times better than for O_2. As a result, the O_2 and CO_2 transfer rates are in practice approximately balanced in this lung.

* Dow Corning

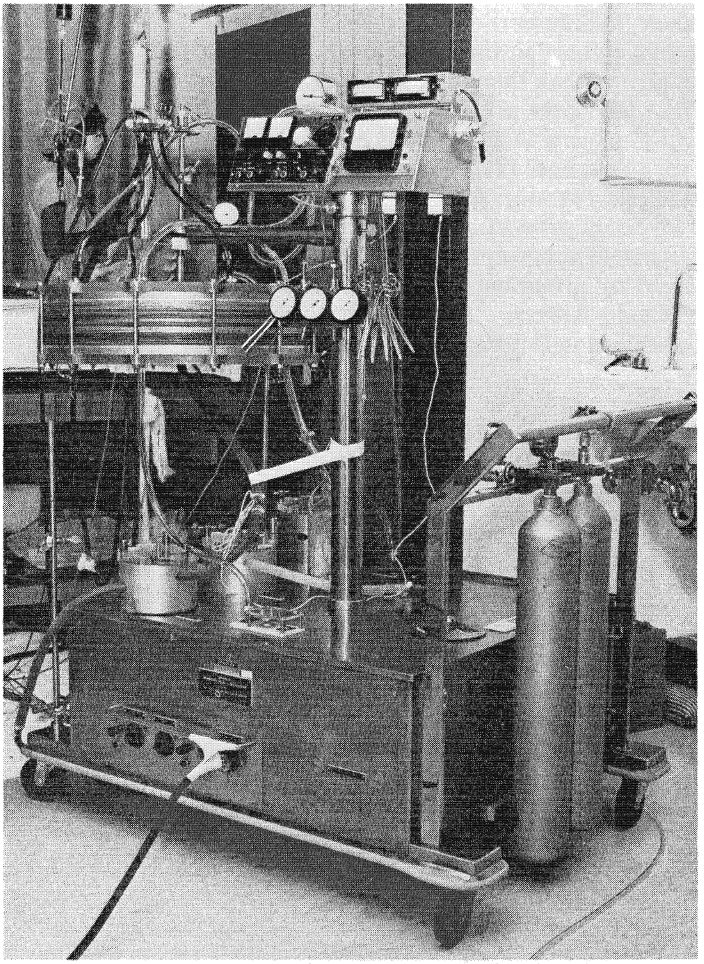

Figure 3.1. The Bramson Membrane Heart–Lung Machine. The 'drum' at the upper left is the membrane lung. (Photo taken in the CPU during prolonged extracorporeal circulation)

Engineering design

Membrane lung design is a problem in fluid logistics (Galletti)[4]

The adult-size lung is a circular drum (*Figure 3.1*) containing 15 'sandwiches' or 'cells'. The stacking order of one cell is shown in

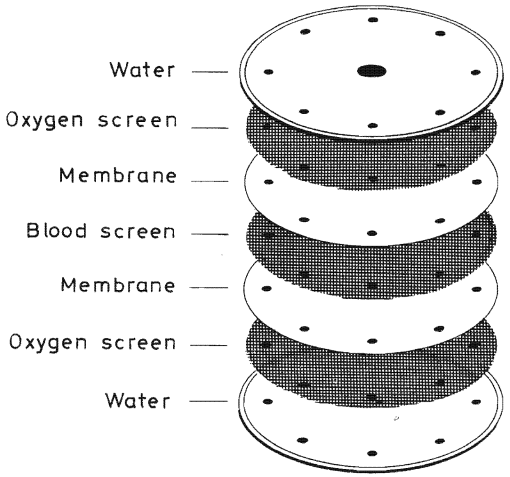

Figure 3.2. The stacking order of one of the 15 'cells' of the membrane lung

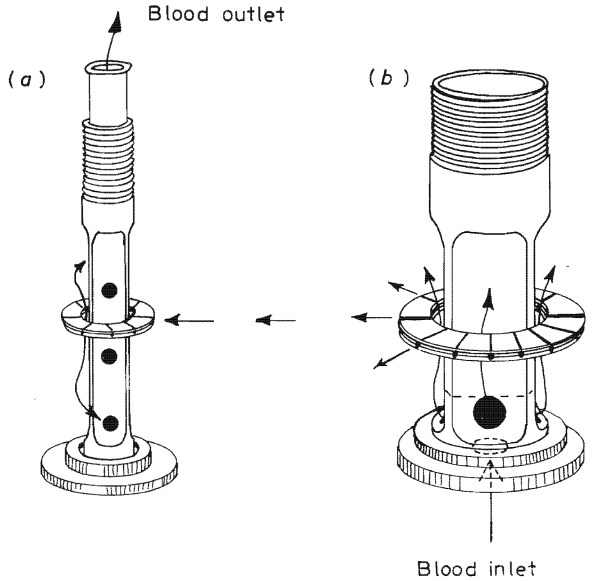

Figure 3.3. (a) Shows one of the eight peripheral manifold bolts. There are four 'blood out' bolts, two water bolts (one 'in' and one 'out') and two oxygen bolts (one 'in' and one 'out'). All these are identical. (b) Shows the central blood inlet manifold bolt with one of the 15 polyethylene distribution washers

Figure 3.2. The thickness of the blood screen (*Figure 3.4a*) defines the blood path between the two membranes which are pressed onto the screen by two pressurized water mattresses (*Figure 3.5*).

Two oxygen screens (*Fig. 3.4b*) are inserted between the membranes and the mattresses, thus giving access of oxygen to each dry membrane surface. The oxygen screens also transmit mechanically the water mattress pressure to the membranes. These pressurized mattresses serve the following purposes: (1) By uniformly supporting the membranes they protect them against bulging and tearing stresses; (2) Provided the water pressure is always set to exceed the blood pressure, they insure constant and minimum blood volume (as defined by the thickness of the blood screens) in the lung; (3) By preventing bulging of the membranes, they compel the blood flow to conform to the blood screen mesh, thus insuring the break-up of the plasma boundary layer by gentle turbulence; (4) By circulating the water and controlling its temperature, integral heat exchange is provided within the lung at no cost in additional equipment or blood contact surfaces.

Manifold systems

The lung is a 'drum' measuring 24 inches in diameter and 6 inches in overall thickness. It contains 15 blood cells (with a total effective membrane area of 6 square metres), 30 oxygen screens and 16 water mattresses. *Figure 3.3a* shows a stainless steel manifold bolt with one distributing ring. The bolt is tubular and externally of square cross-section. Eight such bolts provide gas inlet and outlet, water inlet and outlet and four blood outlets. The central blood inlet bolt is shown in *Figure 3.3b*. It is of larger diameter and also serves as the suspension point for the lung. Blood, gas or water flows through the spaces between the square bolts and the circular holes in the distributing rings and, in parallel, through all of the blood cells, all of the oxygen screens and all of the water mattresses respectively. The distributing rings are injection mouldings, and one nut on each bolt tightens all joints along it.

Figures 3.4a and *3.4b* show the screens and flow patterns for blood and gas.

CIRCUIT AND CONTROLS FOR CARDIAC SURGERY

Blood circuit (*Figure 3.6*)

Venous blood from the patient enters a 'T' connector in the line between the priming reservoir and a slack plastic bladder or 'atrium',

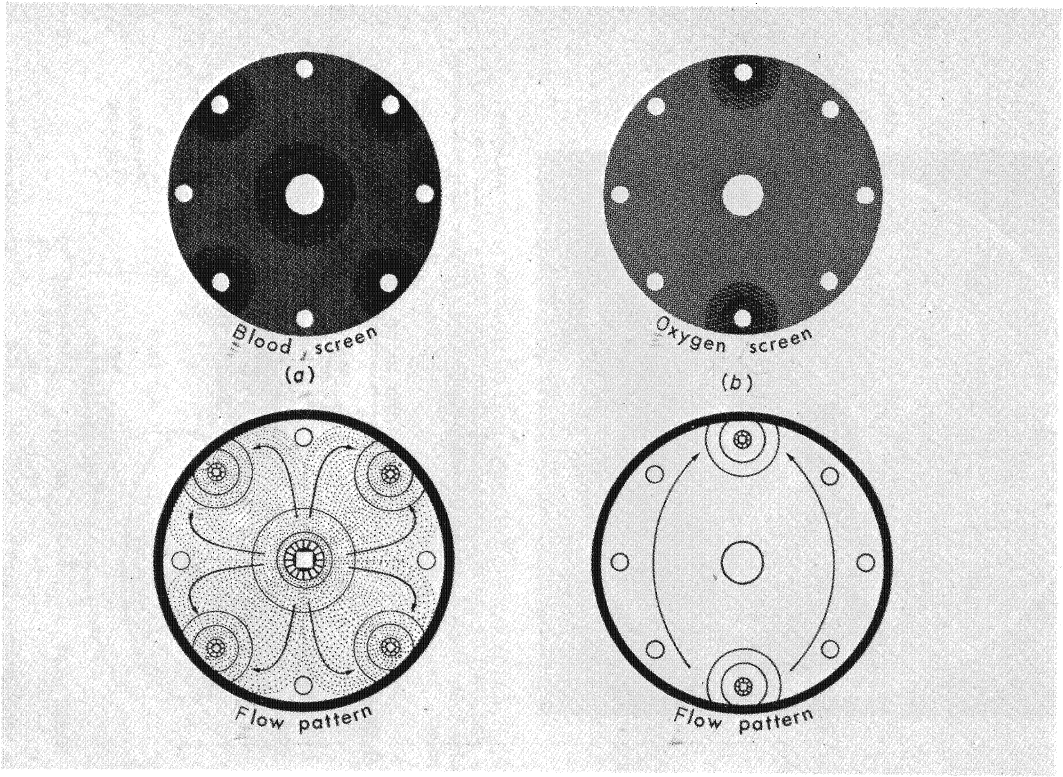

Figure 3.4. (*a*) *Shows a blood screen and the blood flow pattern through each blood screen.* (*b*) *Shows an oxygen screen and the gas flow pattern through it*

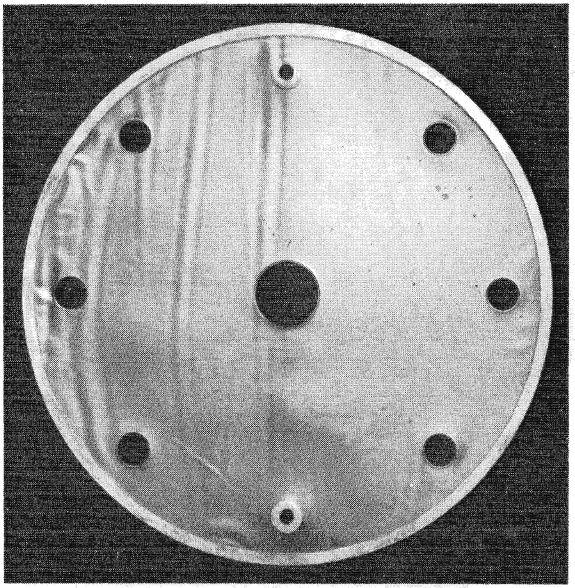

Figure 3.5. A water mattress. The central hole and six of the peripheral holes in the double plastic (PVC) disc are edge-welded. Distribution discs are located at the two remaining holes. When assembled the water manifold bolts (Figure 3.3a) pass through these holes

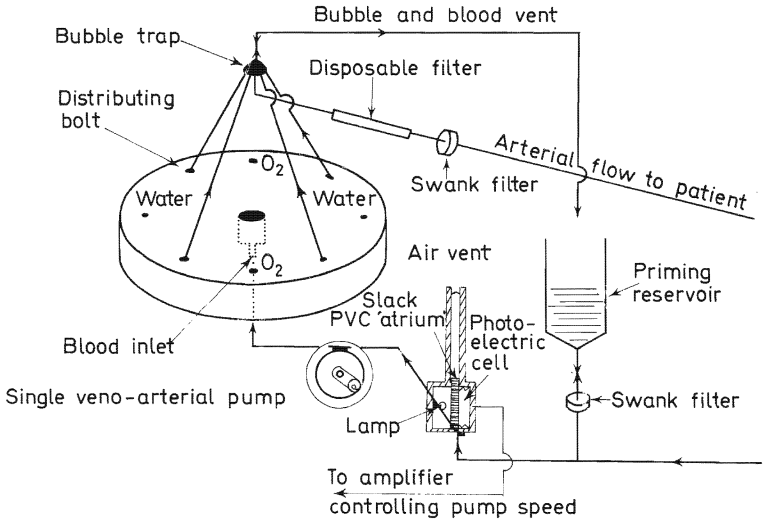

Figure 3.6. The extracorporeal blood circuit

in which the blood level is automatically kept constant. At a lower corner of this bladder a small box is attached which contains a light bulb and a photoelectric cell on opposite sides of the bladder. When the blood level rises it gradually obscures the light reaching the photoelectric cell. This produces a proportional signal which is amplified to control the speed of the arterial pump. (This is the only pump in the circuit and no recirculation is used.) From the 'atrial' bladder the blood passes through the pump into the central blood inlet of the lung, traversing all 15 cells radially in parallel. It emerges from the upper ends of the four blood outlet manifold bolts into four plastic tubes joined together in a disposable bubble trap. Here the stream changes direction and flows downward through a tubular and a 'Swank' filter forming part of the arterial line to the patient.

Two gauges show the lung inlet and outlet pressures and there is an alarm and circuit breaker (defeatable) which prevents a preset line pressure from being exceeded. When desired, such as during priming, the speed control can be switched from automatic to manual.

Oxygen circuit (*Figure 3.7*)

Oxygen enters the lung at one of the oxygen manifold bolts, passes through the 30 oxygen screens (*Figure 3.4b*) also in parallel, and emerges into the atmosphere at the lower end of the diametrically

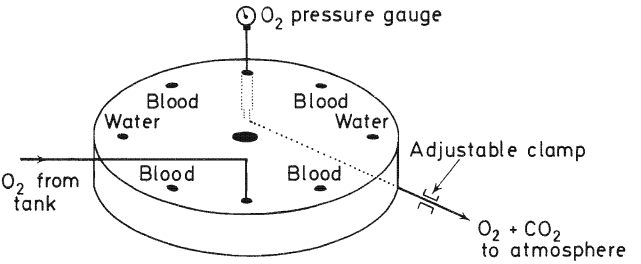

Figure 3.7. The oxygen circuit

opposite oxygen bolt. If it is desired to raise the arterial P_{O_2}, oxygen pressure can be increased by a slight throttling of the gas outflow at this point. To keep the P_{CO_2} in the gas phase low, a large excess of gas flow (about 15 l/min) is used. During total perfusions oxygen with 2 per cent or 4 per cent CO_2 has been found to ensure balanced gas exchange while preventing CO_2 elimination beyond physiologically desirable rates.

Water circuit (*Figure 3.8*)

Water from a tank pressurized from a standard compressed nitrogen bottle is circulated through the water mattresses by a small centrifugal pump and returned to the tank. The water temperature is controlled by means of a coil of copper tubing in the tank through

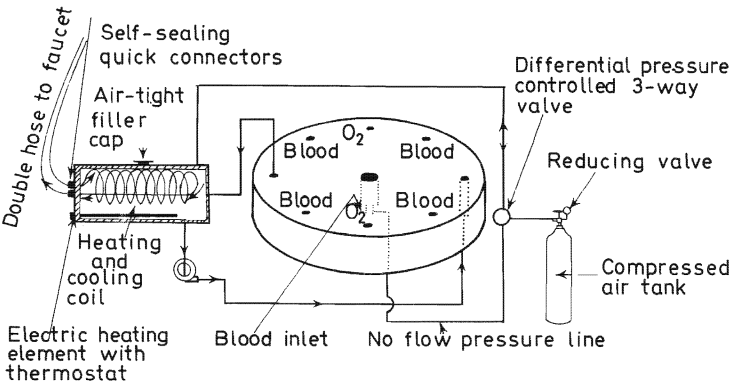

Figure 3.8. The water circuit

which hot or cold water from the nearest faucet can be circulated. (A double rubber hose with self-sealing quick connectors is used for this purpose.) This provides quick cooling if called for, and rewarming. For slow warming a thermostatically controlled electric heater in the tank is used.

The gas pressure in the water tank is controlled by a specially developed three-way valve, which automatically maintains a constant excess over the blood inlet pressure in the lung. The differential is set at about 150 mmHg.

Cardiotomy suction return

The standard chassis comprises three suction roller pumps—one for ventricular venting and the other two for coronary return and general suction from the surgical field.

Modified circuitry for prolonged partial perfusion

The equipment used for prolonged circulatory support of patients in acute pulmonary insufficiency has been slightly modified in the light of experience.

Since heparin must be more closely controlled at lower levels than during short perfusions, it has been found essential to eliminate from the blood circuit all stagnant or semi-stagnant regions. Four such regions were found. Two capillary polyvinyl chloride tubes connecting blood entry and exit to the respective pressure gauges, as well as the gauges, were eliminated and substituted by electronic leads and instrumentation performing the same function.

One capillary polyvinyl chloride tube connecting blood entry with the differential pressure controller was also eliminated. In this case no other instrumentation is needed, since it has been found safe and satisfactory merely to set the water circuit pressurization at a fixed level of 600 mmHg. Lastly, a stagnant region was found in the corner of the 'atrial' bladder where the photoelectric control box was attached. This was redesigned and the stagnant region eliminated.

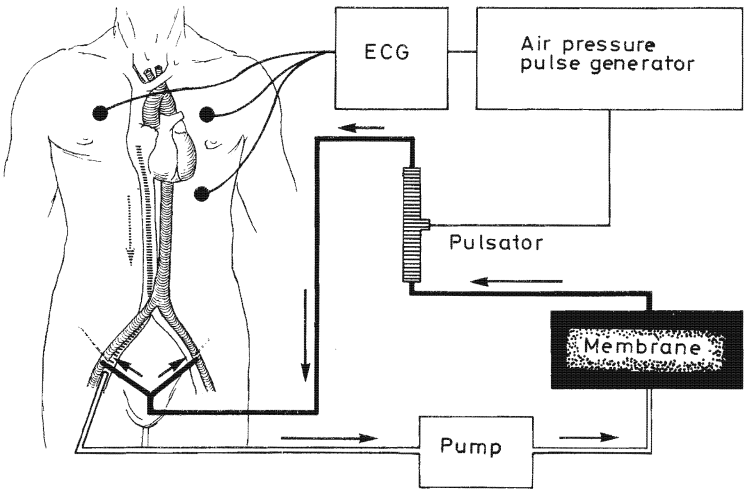

Figure 3.9. Circuit diagram and circulation for membrane lung perfusion combined with diastolic augmentation

PROLONGED PARTIAL PERFUSION WITH DIASTOLIC AUGMENTATION

Circuitry and additional equipment required

Figure 3.9 shows the cannulation and the extracorporeal circuitry required for partial veno-arterial perfusion combined with diastolic

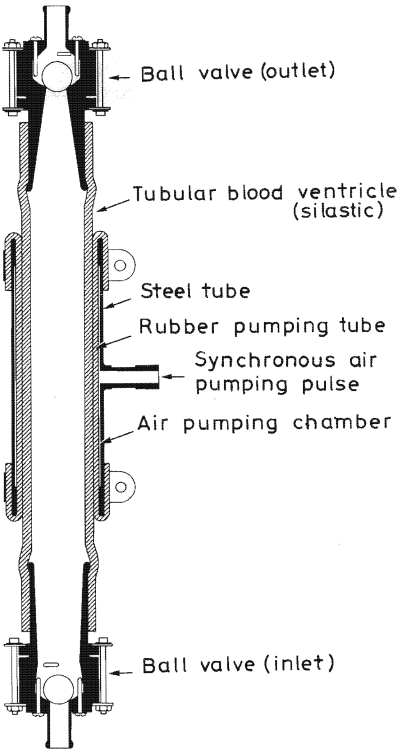

Figure 3.10. Cross-section of tubular ventricle-type pulsator

augmentation to be used for circulatory support in cardiogenic shock.

Figure 3.10 shows a cross-section and *Figure 3.11* an external view of the tubular, ventricle-type pump developed for that purpose. *Figure 3.12* shows the electropneumatic synchronous pressure pulse generator which is triggered by the electrocardiograph. The system

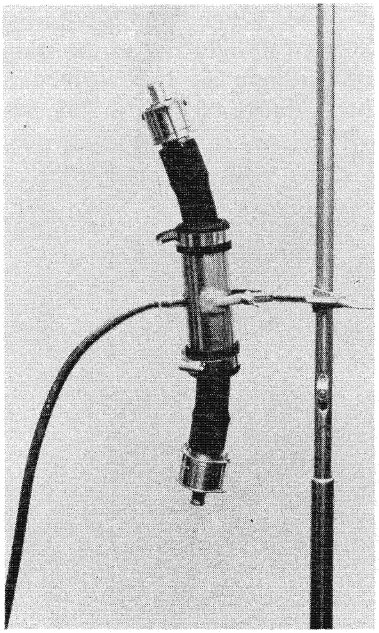

Figure 3.11. External view of tubular ventricle-type pulsator

is so devised that if no pulses arrive at the tubular pump both of its valves remain open, and it simply becomes part of the arterial line.

Clinical performance of the Bramson heart–lung machine

We have reported elsewhere[13] on a series of 22 consecutive clinical perfusions using this equipment. The following is a quotation from that report: '... the longest perfusion lasted 730 minutes and the shortest lasted 28 minutes with an average duration of over two hours. Blood flow through the oxygenator ranged from 2 to 4 litres/min. Mean blood pressure during perfusion was 80 mmHg. The oxygenating capacity of the membrane lung was very satisfactory. The mean P_{O_2} of the arterialized blood was 157 mmHg. The mixed venous P_{O_2} ranged from 31 to 57 mmHg, with a mean of 39 mmHg during perfusion. Carbon dioxide elimination was also excellent. The mixed venous P_{CO_2} was never over 50 mmHg (mean 41.4). The mean P_{CO_2} gradient from the venous to the arterial side of the lung was 6.7 mmHg.

'Metabolic acidosis was never a problem during perfusion. About half of the cases showed a slight rise in the standard bicarbonate; the other half showed a slight fall. The mean change was −0.2 mEq. The arterial pH had a range of 7.32 to 7.63, with a mean of 7.44.

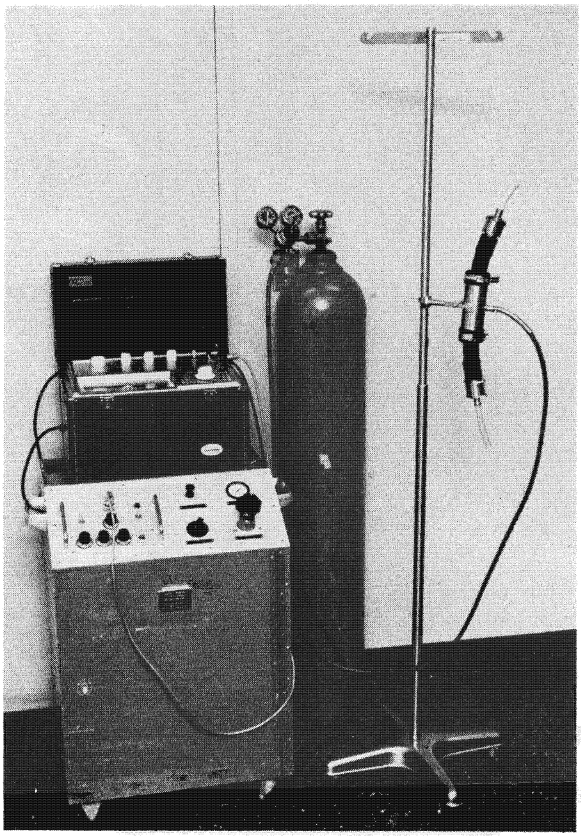

Figure 3.12. Electropneumatic pulse generator, with helium tank, electrocardiograph and pulsator

Urinary flow persisted normally during perfusion in all of the patients in whom it was measured and there was no evidence of post-operative renal damage due to perfusion.

'The mean platelet count at the end of perfusion was measured in 6 patients. In 4 of these patients serial platelet counts were carried out at 5-minute intervals. The mean count at the onset was 220 000 per mm^3; the mean lowest count during perfusion was 91 000

per mm^3, while the mean count at the end of perfusion was 125 000 per mm^3. Thus, there was a tendency for the platelet count to rise during perfusion, a finding we had not observed with our disc oxygenator.

'Haemolysis in the lung was minimal. During *in vitro* recirculation of 2 litres of human blood at 3 litres/min against a pressure of 100 mmHg the rise in plasma haemoglobin was 10 mg/h. During clinical perfusions the amount of haemolysis appeared to be related to the amount of open-heart suction. The rise in plasma haemoglobin ranged from 12 mg per cent/hour to over 40 mg per cent/hour – the higher value being reached when blood was sucked from the pleural space for re-use. It is not possible to measure the haemolysis produced by passage through the lung as separate from that produced by the pump and the arterial cannula, but assuming reasonable values for pump and cannula haemolysis, the amount due to circulation through the lung itself would be under 10 mg per cent per hour of perfusion. The pressure gradient through the lung at a flow rate of 3 litres/min was 15 mmHg at a blood temperature of 32°C. It rose as the blood temperature dropped. At a flow rate of 2.6 litres/min and a blood temperature of 17°C the gradient across the lung was 45 mmHg.

'The heat exchanging capacity of the membrane lung was very large and as a practical matter was limited mainly by the flow and temperature of the water pumped through the spacing mattresses. The temperature of the arterial blood leaving the lung at normal flow rates came very close to reaching the temperature of the circulating water. Because the heat exchange capacity of the lung exceeded what seemed to us to be safe rates of induction of hypothermia or of rewarming, we did not explore how fast the patient could be cooled or rewarmed. Adults were easily cooled or rewarmed at a rate of .5°C per minute ...'

Subsequent clinical experience (the series now exceeds 500 perfusions at this Centre) has shown essentially a similar performance of the system.

Although we have selected for detailed treatment the Bramson membrane lung as exemplifying the four types currently in routine clinical use, it is of interest to tabulate the comparable performance and other important characteristics of all four types. The information has been obtained from published reports, mostly written by the respective designers, and is set forth in Table 3.1.

TABLE 3.1

Characteristics of the Landé–Edwards, General Electric–Peirce, Kolobow and Bramson Membrane Lungs

	Landé–Edwards	*General Electric–Peirce*	*Kolobow*	*Bramson*
Membrane area (m^2)	3.0	3.0	3.5	6.0
Rated flow (l/min)	1.5	1.35	3.5	4.25
Specific rated flow ($l/min/m^2$)	0.5	0.45	1.0	0.71
Oxygen transfer at rated flow (ml/min)	84	78	150	230
Specific oxygen transfer at rated flow ($ml/min/m^2$)	28	26	43	38
Priming volume—lung only (ml)	500	600	350	1100
Specific priming vol.–lung only (ml/m^2)	167	200	100	183
Plasma-free haemoglobin	Remains within control levels	10%/hour *in vitro* at 1 litre/min/m^2 flow	Remains within control levels	Remains within control levels
How is plasma boundary layer controlled	By recirculation	Cones moulded in membranes	Nylon mesh incorporated in membranes	Screens in blood path
Recirculation (%)	200	10	None	None
Constant blood volume	No	No	Yes	Yes
Heat exchange	None	None	External	Yes
Total disposability	Yes	Yes	Yes	No
No. of lungs needed for total perfusion of adults	2 or 3	2	1 or 2 smaller	1
Lung cost per perfusion	$400 or $600	$800	$200 to $350	$250

PERFUSION PROCEDURE FOR CARDIAC SURGERY—OPEN-CHEST TOTAL BODY PERFUSION

Pre-priming

Before allowing any flow into the blood circuit the water circuit must be filled (using distilled or demineralized water) and pressurized (with nitrogen), setting the reducing valve at 15 psi.

The blood circuit may now be filled with heparinized Ringer's lactate (or other preferred electrolyte solution) and flushed two or three times, using the procedure for debubbling as well as rinsing.

For these purposes the recirculating line is used, partially clamped to maintain the pressure around 100 mmHg. Once this pressure is established it should not be allowed to drop, so as not to prolong the debubbling procedure.

To eliminate any danger of air embolism or foaming, it is essential, both during pre-priming and priming, that the circuitry and the procedure be such as to make it impossible for the pressure in the blood space to fall below atmospheric. This can most simply be done by suspending the priming reservoir above the level of the lung.

Priming

When all air has been eliminated, the solution is slowly displaced by 1500 ml of fresh, heparinized donor blood. To increase the fluid reserve an additional 500 ml of heparinized solution is added to the priming reservoir. The surgical sash can now be filled by allowing repeated rapid flush-through from the control bladder via the pump, lung and surgical sash back to the priming reservoir, where any air from the sash is eliminated. At this point or earlier slow warming can be started by switching on the immersion heater in the water circuit.

The arterial line is now clamped, and slow recirculation is started until perfusion is called for. It is good practice, while waiting, to switch the pump control to 'automatic', then controlling the pumping rate by adjusting a screw clamp on the gravity line from the priming reservoir to the control bladder.

Starting perfusion

Starting the perfusion is simply done by the following manoeuvres:

(1) The gravity line from the priming reservoir and the recirculating

line are simultaneously clamped, thus stopping the pump and retaining the pressure in the circuit.

(2) All clamps on the arterial and venous lines, except the large venous screw clamp, are removed.

(3) The latter is slowly opened, thereby starting the perfusion.

When full perfusion is called for it is obtained by fully opening the venous screw clamp. The perfusion rate can be read directly from the pump speed indicator.

GENERAL PERFUSION MANAGEMENT

Adding blood volume

If the patient's central venous or mean arterial pressure or both indicate hypovolaemia, volume may be added by allowing flow from the priming reservoir into the control bladder. This immediately speeds up the arterial pump and, since no 'pooling' can occur in the lung, the exact volume added is at once supplied to the patient.

Removing blood from the patient

If by accident or otherwise the patient is deemed to be hypervolaemic a measured amount of blood can be removed from the circuit by briefly opening the clamp on the recirculating line. It is convenient to have a calibrated volume scale on the priming reservoir.

Increasing the gas exchange capacity of the lung

In cases of exceptionally heavy metabolic oxygen demand, oxygen transfer may be increased by 20–30 per cent by the simple means of slightly throttling the O_2 outflow from the lung. When doing so, the oxygen pressure in the lung must not be allowed to exceed the lowest (i.e. outlet) pressure of the blood, so that even if a pinhole were present in a membrane no gas emboli could occur.

In general, raising the oxygen pressure to 200 mmHg should be safe.

Conversely, since routinely the gas mixture used is 4 per cent CO_2

in 96 per cent O_2, by merely switching to 100 per cent oxygen the P_{CO_2} gradient is increased by 30 mmHg, which will produce a more-than-adequate corresponding increase in the CO_2 transfer capacity.

Heparin dosage

In addition to the initial heparin dose, hourly maintenance of 1 mg of heparin per kg of body weight is given.

Going off perfusion

By slowly closing the venous screw clamp the arterial pump is slowed and stopped. The manual pump speed control should now be set to zero and the switch thereafter changed from 'automatic' to 'manual'. If at this point there is little or no reserve in the priming reservoir, 500 to 1000 ml of heparinized solution should be added, and the clamp between the reservoir and control bladder should be removed. If, as is frequent, transfusion is now required to normalize the patient's blood pressure, any needed proportion of the extracorporeal blood volume can now be delivered from the lung by operating the manual pump control.

Changes in technique for prolonged partial perfusion

Apart from the minor changes in instrumentation referred to above, and in the control of clotting time described in detail in Chapter 12, there are only the following slight changes in set-up and procedure for prolonged perfusion. Since the patients are awake, even though mildly sedated, their metabolic rates and oxygen demand are substantially greater than those of anaesthetized patients undergoing open-heart surgery. The total systemic flow may in a large and young patient amount to 10 litres/min or more. Seventy-five per cent thereof, or approximately 7.5 litres/min may easily be diverted through the artificial lung, which therefore may need a larger capacity. We therefore routinely use a 20-cell lung (instead of the standard 15-cell lung) in the emergency membrane heart–lung machine standing by for prolonged perfusions.

To avoid excessive platelet losses, the Swank filter in the arterial

line is omitted. However, the filter in the line from the priming reservoir is initially retained, so that all donor blood is filtered.

After a few hours of perfusion the priming reservoir and the Swank filter (which contain stagnant regions) are disconnected from the circuit. The distal end of the recirculating line is connected to the control bladder, and a flow of about 50 ml/minute is allowed to prevent stagnation in that line.

The silastic tube of the arterial pump is generously lubricated every 24 hours.

SUMMARY

The design and development of new membrane lungs and the refining and perfecting of those which are already in clinical use is accelerating.

The main stimulus is the recent clinical confirmation of the feasibility of prolonged perfusions with membrane oxygenation as a life-saving procedure in some cases of acute cardiopulmonary insufficiency, otherwise lethal.

Conversely, this trend is further reinforced by the well-established inability of patients to survive perfusions with non-membrane systems lasting more than 5 or 6 hours.

The practical parameters and performance characteristics needed in any heart–lung system for the total perfusion of adults are discussed and the relative advantages and disadvantages of membrane lung systems compared with 'conventional' types are enumerated. On that basis a rationale for using membrane lungs not only for prolonged perfusions, but also for conventional open-heart procedures, is presented.

Of the four membrane lung perfusion systems currently in clinical use, the Bramson membrane heart–lung system is selected for practical detailed description and performance analysis because of the authors' extensive clinical experience with it for open intracardiac surgery, as well as for prolonged perfusions.

Comparable performance and other important characteristics of all four types of membrane lungs are tabulated.

There are now so many other new types of membrane lungs in the course of development that it is difficult to predict a definite trend. Some are of the sandwich type, others of the spiral coil, hollow fibre, oscillating, or rotating disc types.

Many of these are highly efficient with regard to oxygen transfer per minute per square metre. However, when that high efficiency is used to reduce the total membrane area, the CO_2 transfer, which is

'membrane limited', tends to suffer, and is likely ultimately to determine the membrane area and the size of the artificial lung.

Membrane lungs may thus be said to be in a technologically exciting phase, and the sheer volume of new developments reinforces our belief that they hold the future of extracorporeal oxygenation.

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