Strategies for optimisation of paediatric cardiopulmonary bypass

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Chapter 4 Oxygenation by artificial lung systems

The artificial lung or oxygenator is the most technical part of the cardiopulmonary bypass circuit. The design objectives of the “ideal” oxygenator are still the same as in 1962 when Galletti and Brecher [1] described, the “ideal” oxygenator as one that provided: oxygenation of venous blood, carbon dioxide elimination, minimum blood trauma, small priming volume and safety.

Today almost 100% of the oxygenators used are membrane oxygenators [2]. Meaning that a membrane separates the gas and the blood phase. The majority of devices use a microporous hydrophobic membrane. Beside the function of gas exchanger most devices incorporate a heat exchanger and a reservoir. Thus the oxygenator performs all major functions of the natural lungs except for their endocrine function, which can be suspended for a short time without major ill effects.

4.1. The venous reservoir

There are two basic types of venous reservoirs: closed and open. The closed system consists of a PVC bag with an in and outlet and one or more venting ports for the evacuation of air. Advantages of the closed system are almost no blood-air interface, small foreign surface area; collapse of the outlet when the reservoir is suddenly emptied; quick indication of fluid changes and the ability of volume controlled weaning from cardiopulmonary bypass. Disadvantages are a more difficult air removal, when air accidentally enters the system, and the need for an additional cardiotomy reservoir.
The open system is in essence a reservoir open to the atmosphere with incorporated cardiotomy reservoir. This system is somewhat easier to set-up than a closed system and avoids the use of an additional cardiotomy reservoir. When accidentally large amounts of air enter the reservoir, this can be faster removed than in a closed system. The major disadvantages are the large foreign surface area, the hold-up of volume in filter and defoamer and the risk of inadvertently pumping air.

4.2. The heat-exchanger

The working principle of a heat exchanger is based on the principles of conduction and forced convection. Water is used to control the temperature of the blood. A common misconception is that the blood side is the determining factor for performance. The water side is as important because it is desirable to have high flows and turbulent flow to promote conductance. On the blood side it is important to maintain laminar flow to minimise blood component damage, but also to keep the total cross sectional area for blood flow as small as possible to increase conductance [3].

The material used for the separation between the blood and water flows should be as thin as possible for the highest conductance, with a very high thermal conductivity, yet still have the integrity to withstand the expected water and blood side pressures without failure.

Unfortunately, the most haemocompatible materials used in extracorporeal blood handling devices have very poor thermal conductivities (k)(Table 1). There is a trend to use more polymeric heat exchangers since these can be
more easily coated or surface modified compared to metal heat exchangers [3], what makes them more haemocompatible.

Table 1: Thermal conductivity of different materials

<table>
<thead>
<tr>
<th>Material</th>
<th>K  (W/m.K)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stainless steels</td>
<td>15.1</td>
</tr>
<tr>
<td>Aluminium</td>
<td>237</td>
</tr>
<tr>
<td>Polycarbonate</td>
<td>0.2</td>
</tr>
<tr>
<td>Silicone</td>
<td>0.2</td>
</tr>
<tr>
<td>Epoxy</td>
<td>0.2</td>
</tr>
<tr>
<td>Polyurethane</td>
<td>5</td>
</tr>
</tbody>
</table>

4.3. The gas exchanger

The intrinsic physico-chemical and transport characteristics of the membrane, the fluid dynamics and the haemocompatibility of the membrane module will all determine its final mass transfer. As soon as blood gets into contact with the hydrophobic polymeric surface, the material will adsorb proteins. The amount of the protein layer and the nature of the proteins that are adsorbed will depend on the physico-chemical characteristics of the membrane and on the fluid dynamics in the membrane module. Poor fluid dynamics in the blood flow channel of the membrane module will affect dramatically its performance [4] because of:
1. High blood boundary layer resistance to mass transport. This remains extremely important since the resistance to mass transfer in a microporous membrane oxygenator is fluid bound.

2. Poor haemocompatibility. High shear rates, eddy formation and stagnation will favour the occurrence of clotting [5]

3. Large membrane surface. This will is needed for obtaining enough mass transfer but will on the other hand cause activation of the complement system [6,7]

In order to obtain the best possible fluid dynamics, most manufacturers use today extra luminal flow (ELF) designs. In this design blood is flowing outside regularly spaced hollow fibres. The hollow fibres are delivered knitted together in a double layer mat. The membrane module is manufactured by wrapping a double layer hollow fibre mat around a solid core, which is then inserted into a cylindrical shell. In these modules blood flows through the membrane mesh while gas flow is fed counter-currently into the hollow fibres. Since flow through the membrane mesh will be forced to flow partly along and partly around each hollow fibre secondary flows will be generated. This particular membrane arrangement induces mixing in every section of the membrane module to an extent that will depend on the membrane angle with respect to the main direction of blood flow [4]. The efficient destruction of boundary layers by this “static mixer” configuration leads to reduced resistance to mass transfer [8] and yields high transfer rates across the membrane. Aside of the better mass transfer this design has also lower pressure drops at the blood side and no sharp edges in the blood flow path resulting in a better haemocompatibility.
The introduction of this ELF design in paediatric oxygenators has resulted in an important reduction of the total priming volume of the paediatric cardiopulmonary bypass (Figure 1).

**Figure 1. Evolution of priming volume in the University Hospital in Gent**

Nevertheless the priming volume and the total amount of foreign material remains large in even the smallest circuits (Figure 2). Each millilitre of blood in such a small paediatric cardiopulmonary bypass is still exposed to more than three times the amount of foreign surface compared to an adult circuit. This will have a major impact on the inflammatory response [7].

**Figure 2. Relationship between blood volume and foreign surface**
4.4. Fluid dynamics and shear stress

As pointed out when describing the gas exchanger module, fluid dynamics is an important item for obtaining optimal mass transfer. However an oxygenator does consist of different components which must be connected. At the same time, blood has to be evenly distributed over the heat exchanger and through the membrane mesh by manifolds. As a result blood velocity will change when blood passes through the oxygenator and this may result in zones of stasis, eddy formation and or high shear. The average shear stress at the wall in a membrane oxygenator can be calculated by starting with the general macroscopic force balance for flow in a tube [9]. However, tube flow does not accurately represent the complex flow in an ELF oxygenator. Flow through an oxygenator can be considered as flow through a porous medium. According to Bird [10] the shear stress in each oxygenator was calculated by considering the flow equivalent to the flow in a packed column governed by:

\[ \tau = \frac{R_h \Delta P}{L} \]

where: \( \Delta P \) = pressure drop \([N/m^2]\), \( L \) = blood path length \([cm]\), \( R_h \) = hydraulic radius \([cm]\)

\[ R_h = \sqrt{\frac{Q(25/6)\mu L}{\Delta P \varepsilon (A e)}} \]

where: \( \varepsilon \) = porosity of membrane area that fills that cross section

\[ Q = \text{volumetric pump flow} \ [L/min] \]
\[ \mu = \text{dynamic fluid viscosity} \ [N/m².s] \]
\[ A_e = \text{cross sectional area for flow} \ [m²] \]
\[ 25/6 = \text{experimental derived factor.} \]
Mockros proposed a different formula for calculating shear in an oxygenator \([11,12]\).

\[
\tau = \left[ \frac{\Delta P Q \mu}{V} \right]^{\frac{1}{2}}
\]

where: \(V = \) volume oxygenator \([L]\)

The average shear for two different neonatal oxygenators calculated by both formulas are given in table 2.

**Table 2: Characteristics of two neonatal oxygenators**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Dideco D901</th>
<th>Polystan Safe Micro</th>
</tr>
</thead>
<tbody>
<tr>
<td>Membrane Surface area, m²</td>
<td>0.34</td>
<td>0.33</td>
</tr>
<tr>
<td>Heat Exchanger Area, m²</td>
<td>0.02</td>
<td>0.05</td>
</tr>
<tr>
<td>Void volume</td>
<td>0.58</td>
<td>0.48</td>
</tr>
<tr>
<td>Total priming volume, cm³</td>
<td>60</td>
<td>52</td>
</tr>
<tr>
<td>Blood Pressure Drop @ 0.8 lpm (mmHg)</td>
<td>95</td>
<td>51</td>
</tr>
<tr>
<td>Blood Pressure Drop @ 0.6 lpm (mmHg)</td>
<td>65</td>
<td>35</td>
</tr>
<tr>
<td>Blood Pressure Drop @ 0.4 lpm (mmHg)</td>
<td>40</td>
<td>21</td>
</tr>
<tr>
<td>Blood Pressure Drop @ 0.2 lpm (mmHg)</td>
<td>20</td>
<td>9</td>
</tr>
<tr>
<td>Blood Path Length oxygenator, cm</td>
<td>30</td>
<td>15.3</td>
</tr>
<tr>
<td>Average Cross sectional area for flow, cm²</td>
<td>12</td>
<td>7.62</td>
</tr>
<tr>
<td>(\tau) oxygenator (Ben Brian) [dynes/cm²]²</td>
<td>18</td>
<td>17</td>
</tr>
<tr>
<td>(\tau) oxygenator (Mockros) [dynes/cm²]²</td>
<td>25</td>
<td>20</td>
</tr>
<tr>
<td>(\tau) membrane compartment (Ben Brian) [dynes/cm²]²</td>
<td>31</td>
<td>19</td>
</tr>
<tr>
<td>(\tau) membrane compartment (Mockros) [dynes/cm²]²</td>
<td>7</td>
<td>5</td>
</tr>
</tbody>
</table>

Although the calculated values are comparable with those in blood vessels (see chapter 2), these values are average values and do not exclude that at certain points in the design shear stress is above the critical level of 75 – 100 dynes/cm² needed to activate white blood cells and platelets.

\[1 \text{ dyn}_{\text{cm}^2} = \frac{10^6}{m^2} \text{ N} \]
Every extracorporeal device will have a flow window with “ideal shear”. If shear is too high platelets and blood elements will be damaged but when shear is too low platelets will be more easily adsorbed by the material. As explained earlier not only the magnitude of shear stress is important but also the exposure time to this absolute value. It is well known that high shear for a short time period is better tolerated than average shear during a long exposure time [13]. In order to define spots with high or very low shear stress in a design computational fluid dynamics are used [14-16].

4.5. Conclusions

Major improvements in oxygenator design has led to a large reduction in foreign surface area, better haemocompatibility and enhanced mass transfer. Although fluid dynamics have improved more work should be done to locate risk zones at micro level. Computational fluid dynamics might offer the tool for obtaining this goal. Finally this may lead to the ideal paediatric oxygenator that will combine optimal fluid dynamics and thus mass transfer with a small priming volume and foreign surface area.

References


