The Neonatal Circuit: In Search of the Ultimate Solution

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Abstract: Continuous improvement of both perfusion techniques and perfusion equipment has led to decreased morbidity after neonatal and pediatric cardiopulmonary bypass. Small pediatric circuits have been developed to decrease priming volume and foreign surface area. However, it is not clear if our existing technology allows for further miniaturization of the circuit without jeopardizing safety and hemocompatibility. Keywords: pediatric cardiopulmonary bypass, mass transfer, miniaturization, hemolysis. JECT. 2012;44:P48–P50

Since the first repair of a congenital heart defect with the use of cardiopulmonary bypass (CPB), congenital heart surgery has undergone a huge evolution. As a direct consequence, today more neonates and immature children are put on CPB as witnessed by an increase of the number of children with a birth weight below 8 kg in whom CPB is performed (1). The latter poses new challenges for both industry and the perfusion community and has initiated a continuous search for the most optimal neonatal CPB system. This ideal system should combine low foreign surface area with reliable mass transfer, low priming volume, and optimal hemocompatibility. Over the years, several publications describe very small neonatal CPB systems (2–6). The ingenuity of some of these systems even allows performing procedures without homologous blood products (6).

However, some questions arise. Is it possible to combine all of these features to get the best of all worlds? What about safety? What about when something unexpected happens during the procedure? At what point does downsizing of a system become critical and the benefits no longer balance the possible hazards?

MINIATURIZATION

To obtain an “ideal” neonatal CPB system, one depends for a large part on industry providing small oxygenators, reservoirs, and filters; however, the perfusionist can influence total priming volume by choosing oxygenator position, use of vacuum-assisted return, and the correct choice of diameter and length of tubing. The impact of the latter is shown in Table 1. It is important not to forget that tubing size and length have not only an impact on the priming volume and thus the circulating volume, but even more important on the volume, which is temporarily taken out of the circuit for example by a ventricle vent. In a neonate of 2 kg with a total blood volume of 170 mL, a vent line of 150 cm with a diameter of 3/16-inch, filled with blood, will remove 27 mL blood or 16% of the circulating volume. In many cases, the perfusionist needs to compensate for this volume loss by adding volume while being confronted with excess volume when the vent line is removed.

 MASS TRANSFER

Mass transfer in neonatal oxygenators is important because part of the neonatal population consists of cyanotic children. To address this fact, most oxygenators have oxygen transfer, which is significantly higher than the Association for the Advancement of Medical Instrumentation standards (7). Nevertheless, important differences in working conditions can be found between commercial oxygenators (Table 2). Based on these data, we can calculate the oxygen transfer:

\[
O_2 \text{ Transfer} = (1.34 \cdot \text{Hb} \cdot (S_aO_2 - S_vO_2) + k \cdot (P_aO_2 - P_vO_2)) \cdot Q_b (1)
\]

where: Hb = hemoglobin concentration; 1.34 binding constant hemoglobin; S_aO_2 = arterial oxygen saturation;
\[ \tau = \left( \frac{\eta \cdot Q_b \cdot \Delta P}{V} \right)^{1/2} \]  

where: \( \tau \) = shear stress; \( \eta \) = dynamic viscosity; \( Q_b \) = blood flow; \( \Delta P \) = pressure drop over the oxygenator; and \( V \) = priming volume oxygenator.

It is interesting to note the relationship between oxygen transfer and shear stress. The higher the shear stress, the higher the mass transfer of the oxygenator (Figure 1). However, there is also a correlation between shear stress and hemolysis. Does this implement that an oxygenator with a higher mass transfer will cause more hemolysis? Not exactly because hemolysis will depend on the absolute shear stress, but also on the time blood elements are exposed to that absolute value (8–10). If we calculate the shear stress, exposure time, and hemolysis data again, the differences between oxygenators are small and independent from pressure drop (Figure 2). We also notice that all oxygenators generate some platelet activation at their maximum flow. The latter implies that most likely each device has an optimal working range. However, this optimal range is not necessarily the range given by the manufacturer.

### HEMOLYSIS

Hemolysis is a concern in neonatal CPB because most neonates have low plasma haptoglobin concentrations. Haptoglobin binds free plasma hemoglobin and protects the body against the negative effects of free plasma hemoglobin. Especially the binding between free plasma hemoglobin and nitric oxide is of concern because this will induce pulmonary hypertension (11).

If packed cells are used in the priming, this will further increase free plasma hemoglobin levels. Avoiding packed red cells by maximal reduction of the priming volume seems attractive in this setting. As explained previously, reduction of tubing length and diameter will be beneficial. However, what would be the maximum flow for a given tubing size? Unfortunately, there are no clear guidelines on this so we need to start from a theoretical approach. If
hematocrit and blood temperature remain constant, the following observations can be made:

(1) Based on the equation of energy dissipation,

$$E = \Delta P \cdot \frac{Q_b}{V}$$

where: $E =$ energy dissipated; $\Delta P =$ pressure drop; and $V =$ blood volume a lower blood volume in a tube will result in more energy dissipation to the blood and thus more hemolysis (9,12).

(2) Based on the Poiseuille equation,

$$\Delta P = \frac{128 \cdot \eta \cdot L \cdot Q_b}{\pi \cdot D^4}$$

where: $\Delta P =$ pressure drop; $\eta =$ dynamic viscosity; $L =$ length of tubing; $Q_b =$ blood flow; and $D =$ diameter of tubing:

- Increasing length will give an inverse linear decrease in flow; and
- Decreasing diameter will increase pressure drop by a factor of 5.

Table 3 illustrates the impact of a change in tubing diameter with no change in blood flow. If the desired flow is 1000 mL/min, decreasing tubing diameter from ¼-inch to 3/16-inch will decrease volume by 50% but will increase pressure drop by a factor of 3 and hemolysis by a factor of 8. So one should carefully balance all changes before deciding which tubing diameter is the best option for a given procedure.

**CONCLUSIONS**

Miniaturization of the neonatal CPB circuit has improved the outcome of neonatal cardiac surgery. However, we are reaching the physical limits of the existing technology and further improvements most likely will ask for new technology.

**REFERENCES**


**Table 3.** Impact of changing tubing size on pressure drop, shear stress, velocity, and hemolysis.

<table>
<thead>
<tr>
<th>Diameter (inch)</th>
<th>Diam (mmHg)</th>
<th>V (cm/sec)</th>
<th>Re</th>
<th>$\tau$ (dyne/cm²)</th>
<th>Hemolysis (mg/mL cells*min)</th>
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$\Delta P,$ pressure drop; $V,$ velocity; $Re,$ Reynolds number; $\tau,$ wall shear stress; $Re,$ Reynolds number; hemolysis calculated according to Bluestein and Mockros (9).