Strategies for optimisation of paediatric cardiopulmonary bypass
Somer, Filip Maria Jan Jozef De

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Chapter 2 Vascular access for total body perfusion

2.1. Introduction

This chapter introduces the limitations and boundary conditions of vascular access in paediatric cardiopulmonary bypass. The different requirements for venous and arterial access are reviewed. Finally, the hydrodynamic characteristics and different evaluation methods are presented and discussed. Recommendations for an optimal communication between manufacturer and clinician are given.

2.1.1. Problems related to vascular access

Unsuccessful cannulation may lead to cerebral complications [1-3] A malpositioned aortic cannula may obstruct cerebral blood flow, or it may cause a preferential flow into the descending aorta and “steal” blood from the brain’s circulation [3]. Alternatively, obstruction by the superior vena caval cannula may decrease cerebral venous drainage and potentially lead to brain dysfunction [3]. A direct correlation between age and cerebral alterations (low cerebral blood flow velocity and EEG slowing) caused by malpositioning of the cannulas has been reported [3].

2.2. Venous access

Cannulation of the venous side of the circulation aims at draining the venous blood from the central veins or right heart cavities in a laminar flow without inducing any marked change of the pressure within the large veins. Only then an adequate forward flow can be established. The entire venous return to the
heart should be able to pass through the chosen cannulas without obstruction and without damaging the blood vessel [4].

An essential problem of venous drainage is a compliance and geometric mismatch. Wide, low-resistance, collapsible vessels are connected to smaller, less compliant, artificial conduits. When suction is applied to the venous reservoir, flow starts to increase linearly, but once the vessel starts to collapse, the flow will stagnate. Increase in suction force beyond a critical level, therefore, cannot increase the amount of venous drainage. Additionally, high resistance in the drainage tube necessitates higher degrees of suction than is needed with short, wide tubing. Maintenance of a positive pressure at the tip of the cannula broadens the range of flow regulation because it prevents venous collapse [5]. Reduced venous drainage may be due to reduced venous pressure, inadequate height of the patient above the venous reservoir, malposition of the venous cannulas or obstruction or excess resistance of the lines and cannulas. Venodilation or hypovolaemia may cause inadequate venous pressure.

2.3. Arterial access

Cannulation of the arterial side of the circulation must provide an adequate forward flow of blood to the patient. The cannula and its placement must not be non-obstructive and flow must be directed to the distal aorta in order to perfuse all areas of the body.

The ideal cannula will generate sufficient flow without obstructing or damaging the blood vessel.
2.4. Cannula characteristics

2.4.1. Design related problems

The choice of the best cannula for a given procedure is not simple. In general, manufacturers do not mention in their information brochures the internal diameter of a cannula but only the outer diameter. Depending on the production process, the wall thickness of comparable cannulas can be quite different although their respective manufacturers measured identical outer diameters [6]. Additionally, production tolerances result in important differences in internal diameter even between cannulas of identical size and manufactured by the same company. Since the pressure-flow relation highly depends on the inner diameter and cannulas standard used in paediatric cardiopulmonary bypass have small diameters, this results in significant deviations of the mean values given by the manufacturer.

Another difficulty is related to the fact that the pressure-flow characteristic of a cannula is always measured for water (low viscosity and Newtonian fluid). Unfortunately, it is difficult to extrapolate water values towards blood (higher viscosity and non Newtonian fluid) flow conditions.

2.4.2. Available data for clinicians

Manufacturers only report the polynomial regression of the water data of a certain number of cannulas (Figure 1). Thus, the user has no information about of the possible variability range. This is demonstrated in Figure 1 where both the polynomial regression (full line) as given by the manufacturer and the measured data of ten cannulas (dots) are depicted.
2.4.3. Theoretical relationship

For a horizontal straight tube the relation between pressure and flow can be described by Poiseuille's formula:

\[ \Delta P = \left( \frac{8\mu L}{\pi R^4} \right) Q \]

\[ \Delta P = \left( \frac{32\mu L}{D^2} \right) U \]

where \( \mu \) = dynamic viscosity [N/m².s], \( L \) = length [m], \( R \) = radius [m], \( Q \) = average flow [m³/s], \( U \) = mean velocity [m/s], \( D \) = diameter [m].

For cannulas this formula cannot be used since most cannula are not straight tubes.

2.4.4. Practical characterisation

Several attempts have been described to predict the clinical performance of cannulas.
(1) Montoya et al. propose a system in which any vascular access device can be characterised by a single number denoted as “M” which may be determined from the geometry and/or from simple in vitro pressure-flow measurements [7-9]. M is defined as \( \log (LD_c^{-4.75}) \) where \( L \) represents the length and \( D_c \) the characteristic diameter of the cannula. The \( D_c \) is also known as hydraulic diameter for non-circular ducts representing the diameter of a corresponding circular orifice. The method can be used to choose the best possible cannula when a given diameter or pressure may not be exceeded during the procedure.

Unfortunately, the method has some disadvantages. In order to obtain the M-number on a non-uniform design, such as a cannula, one has to do \textit{in vitro} measurements. The M-number also assumes that the flow regimen is turbulent. However the obtained value is not useable in clinical practice, especially if it is obtained by water measurements. Water measurements tend to lie in the turbulent region while the blood flows used during clinical use are in the laminar region. The latter limits its use in open-heart surgery [10].

(2) Another approach is based on the theory of dynamic similarity [6,11-12]. Flows become identical if the Reynolds number, a measure of the ratio between inertial and viscous forces, is identical for both fluids [6] in the experimental set-up (e.g. water) and in the clinical situation (blood).

\[
\text{Re} = \frac{UD}{v} = \frac{4\rho Q}{\mu \pi D} \quad \text{with} \quad v = \frac{\mu}{\rho}
\]

Where \( Q = \text{flow} \ [m^3/s] \), \( \rho = \text{density} \ [kg/m^3] \), \( \mu = \text{dynamic viscosity} \ [N/m^2 \text{ s}] \), \( D = \text{diameter} \ [m] \), \( v = \text{kinematic viscosity} \ [m^2/s] \), \( U = \text{mean velocity} \ [m/s] \).
For \( \text{Re}_{\text{blood}} = \text{Re}_{\text{water}} \):

\[
Q_{\text{blood}} = Q_{\text{water}} \frac{V_{\text{blood}}}{V_{\text{water}}}
\]

The pressures for a given water flow can be transformed to those of blood in an analogue way by using the Euler number, a measure of the ratio between pressure and inertial forces:

\[
E_u = \frac{P}{\rho U^2} = \frac{\pi^2 D^4 \Delta P}{16 \rho Q^2}
\]

Where \( P = \) pressure [Pa]

For \( E_{u_{\text{blood}}} = E_{u_{\text{water}}} \):

\[
P_{\text{blood}} = P_{\text{water}} \frac{\rho_{\text{blood}}}{\rho_{\text{water}}} \left( \frac{U_{\text{blood}}}{U_{\text{water}}} \right)^2
\]

so that \( P_{\text{blood}} = P_{\text{water}} \frac{\rho_{\text{blood}}}{\rho_{\text{water}}} \left( \frac{\mu_{\text{blood}}}{\mu_{\text{water}}} \right)^2 \)

The dimensionless numbers Reynolds and Euler are independent of the fluid physical properties. This allows converting directly flow rates and pressures. In order to apply this technique one has to know the rate of the densities and the rate of the dynamic and kinematic viscosity of both fluids. Since water tests are performed at room temperature water density is approximately 1000 kg/m\(^3\) (998.2019 kg/m\(^3\)) and water kinematic viscosity 1 \( 10^{-6} \) m\(^2\)/s (1.0038 \( 10^{-6} \) m\(^2\)/s).

If we compare water data with blood at a temperature of 37\(^\circ\)C and a haematocrit of 33.5\% we obtain the following pressure and flow conversion factors presented in Table 1. The factors in table 1 are calculated using the formulas presented in section 3.1.2.3.
Table 1. Pressure and flow conversion factors

<table>
<thead>
<tr>
<th></th>
<th>$Q_{\text{blood}}/Q_{\text{water}}$</th>
<th>$P_{\text{blood}}/P_{\text{water}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T = 37^\circ\text{C}$</td>
<td>2.43</td>
<td>6.21</td>
</tr>
<tr>
<td>$T = 20^\circ\text{C}$</td>
<td>3.40</td>
<td>12.19</td>
</tr>
</tbody>
</table>

Flows and pressures measured during water tests are multiplied with these factors to obtain corresponding blood flows and pressures.

(3) A third method rescales the coefficients of the fitted parabolic equation between pressure drop ($\Delta P$) and flow rate ($Q$)

$$\Delta P_{\text{water}} = a_{\text{water}}Q_{\text{water}}^2 + b_{\text{water}}Q_{\text{water}}$$

to blood

$$\Delta P_{\text{blood}} = a_{\text{blood}}Q_{\text{blood}}^2 + b_{\text{blood}}Q_{\text{blood}}$$

For a given $a_{\text{water}}$, $b_{\text{water}}$ and the relationship between pressure and flow one can determine $a_{\text{blood}}$ and $b_{\text{blood}}$ as:

$$a_{\text{blood}} = \frac{\rho_{\text{blood}}}{\rho_{\text{water}}} a_{\text{water}}$$

$$b_{\text{blood}} = \frac{\mu_{\text{blood}}}{\mu_{\text{water}}} b_{\text{water}}$$

Table 2. Conversion factors for coefficients $a$ and $b$

<table>
<thead>
<tr>
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<th>$a_{\text{blood}}/a_{\text{water}}$</th>
<th>$b_{\text{blood}}/b_{\text{water}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T = 37^\circ\text{C}$</td>
<td>1.055</td>
<td>2.56</td>
</tr>
<tr>
<td>$T = 20^\circ\text{C}$</td>
<td>1.055</td>
<td>3.59</td>
</tr>
</tbody>
</table>

The factors in Table 2 are derived from Table 1 taking into account $a_{\text{blood}} = \frac{\rho_{\text{blood}}}{\rho_{\text{water}}} a_{\text{water}}$.
ratio of 1.03.

In Figure 3 a comparison of both methods (calculation based on dynamic similarity and the parabolic method) is presented. There is still a deviation from the measured data but it gives an estimate of what can be expected under given conditions. The deviation is due to the low accuracy of water measurements caused by the error range on pressure transducers and flow meters. These errors are subsequently multiplied with the conversion factors resulting in even larger deviations. This also explains why the deviation of the calculated data is smaller at 37°C than at 20°C. Use of water-glycerin solutions by manufacturers for validation of their cannulas instead of water will reduce the error.
2.4.5. Quantification of blood damage

Pressure-flow relationships do not give direct information regarding the possible damage of blood elements when a given cannula is used. It is not necessarily the cannula with the highest pressure drop that will generate most damage. The exerted shear rate and specifically shear stress in combination with the duration of these forces (residence time) are far more important factors for blood cell damage [13]. Shear stress equals fluid dynamic viscosity multiplied by shear rate.

\[ \tau = \mu \frac{\delta u}{\delta r} \]

with \( u \) the axial velocity component and \( r \) the radial variable

or

\[ \tau_w = \Delta P \frac{R}{2L} \]

where \( \tau_w \) = shear stress [N/m²], \( R \) = radius [m], \( L \) = length [m]

As tube length is usually several orders of magnitude greater than radius, pressure is generally orders of magnitude greater than shear stress [14]. Physiological values of shear stress range from 1 – 50 dynes/cm² [14]. Most actual cannulas will easily generate shear stresses of several hundred dynes/cm² [15], which is far above the trigger values of 75 and 100 dynes/cm² [14,16] needed to activate white blood cells and platelets, respectively.

\[ 10 \text{ dynes/cm}^2 = 1 \text{ N/m}^2 \]
2.5. Conclusions

Vascular access in neonates and small infants remains a major challenge for adequate paediatric cardiopulmonary bypass. Small vascular structures, congenital malformations and technical limitations in the manufacturing of cannulas give rise to specific problems. A better documentation of the pressure-flow relationship of a cannula in combination with its shear stress data will help the clinician in choosing the best cannula for a given procedure. Thus manufacturers should provide more adequate information regarding the pressure-flow characteristics and both the inner and outer diameter of their products.

References


