A Multiscale Model of the Neonatal Circulatory System Following Hybrid Norwood Palliation

Andres Ceballos*
Alain J. Kassab*, Eduardo A. Divo*, Rubén D. Osorio*
Ricardo Argueta-Morales& and William M. DeCampli&,**

*Department of Mechanical, Materials and Aerospace Engineering
**College of Medicine
University of Central Florida
4000 Central Florida Blvd., Orlando, FL, 32816, USA

*School of Engineering Technology, Daytona State College
1770 Technology Blvd., Daytona Beach, FL 32117, USA

&Congenital Heart Institute at Arnold Palmer Hospital, Arnold Palmer Hospital for Children
92 West Miller St., Orlando, FL, 32806, USA

Abstract

Hypoplastic left heart syndrome (HLHS) is a complex cardiac malformation in neonates suffering from congenital heart disease and occurs in 1 per 5000 births [1]. HLHS is uniformly fatal within the first hours or days after birth as the severely malformed anatomies of the left ventricle, mitral and aortic valves, and ascending aorta are not compatible with life. The regularly implemented treatment, the Norwood operation, is a complex open heart procedure that attempts to establish univentricular circulation by removing the atrial septum (communicating the right and left ventricle), reconstructing the malformed aortic arch, and connecting the main pulmonary artery into the reconstructed arch to allow direct perfusion from the right ventricle into the systemic circulation. A relatively new treatment being utilized, the Hybrid Norwood procedure, involves a less invasive strategy to establish univentricular circulation that avoids a cardiopulmonary bypass (heart-lung machine), deliberate cardiac arrest, and circulatory arrest of the patient during the procedure. The resulting systemic-pulmonary circulation is unconventional; blood is pumped simultaneously and in parallel to the systemic and pulmonary arteries after the procedure. Cardiac surgeons are deeply interested in understanding the global and local hemodynamics of this anatomical configuration. To this end, a multiscale model of the entire circulatory system was developed utilizing an electrical circuit model analogy for the peripheral or distal circulation coupled with a 3D Computational Fluid Dynamics (CFD) model to understand the local hemodynamics. The electrical circuit model is mainly a closed loop circuit comprised of RLC compartments that model the viscous drag, flow inertia, and compliance of the different arterial and venous beds in the body. A system of 32 first-order differential equations is formulated and solved for the electrical circuit model using a fourth-order adaptive Runge-Kutta solver. The output pressure and flow waveforms obtained from the electric circuit model are imposed as boundary conditions on the 3D CFD model. The CFD model domain is a representative HLHS anatomy of an infant after undergoing the Hybrid Norwood procedure and is comprised of the neo-aorta, pulmonary roots, aortic arch with branching arteries, and pulmonary arteries. The flow field is solved over several cardiac cycles using an implicit-unsteady RANS equation solver with the K-Epsilon turbulence model.
Introduction

Congenital heart disease is the leading cause of death for infants born with birth defects, with more than a 30% mortality attributed to this disease. One of the most complex forms of congenital heart disease is HLHS and occurs in 4 to 8 percent of infants born with cardiac malformations [2]. The main complications of HLHS are severely malformed anatomes of the left ventricle, mitral and aortic valves, and ascending aorta (Figure 1). There are currently two approaches to therapy for children with HLHS: heart transplantation or surgical intervention to establish univentricular circulation. Unfortunately heart transplantation is often not an alternative due to the low number of donors at this early age [3]. Surgical techniques such as the Norwood operation are then the most common method to establish an early blood circulation that is compatible with life. The Norwood operation consists of removal of the atrial septum, reconstruction the malformed aorta, and connection of a bypass (BT-shunt) from the innominate artery to the right pulmonary artery (RPA) to allow for pulmonary perfusion (Figure 1). Successful recovery of the patient from the Norwood operation is often followed by subsequent surgical procedures to establish a more normal circulatory system where pulmonary perfusion is achieved with deoxygenated blood. A relatively new alternative to the first-stage Norwood procedure is the Hybrid Norwood procedure, which consists of removal of the atrial septum, implantation of a stent in the patent ductus arteriosis, and banding of the pulmonary arteries [7]. The PDA, a small vessel that connects the pulmonary arteries with the aortic arch and normally closes shortly after birth, is enlarged with the use of the stent to allow circulation from the right ventricle into the systemic circulation through the aortic arch. The pulmonary banding is put in place to prevent excessive pulmonary circulation, with an ideal adjustment providing a pulmonary to systemic flow ratio of 1 [8]. The Hybrid Norwood is a less invasive procedure that avoids the use of a heart-lung machine as well as deliberate cardiac and circulatory arrest. Though there are numerous surgical advantages of the Hybrid versus the conventional method, there are also different complications that may arise with this method. The most important to surgeons participating in this study is the obstruction of flow due to stenosis of the aortic arch proximal to the stent. This condition is fatal if coronary and/or carotid flow is critically low. With the expectation of increasing perfusion to these vessels, a bypass called Reverse BT-shunt (RBTS) is placed connecting the pulmonary root to the innominate artery (Figure 2). The resulting hemodynamics are very complex since the flow reaching the innominate artery is both antegrade and retrograde, meaning part of the circulation flows downstream through the innominate and part upstream to feed the coronary and remaining aortic branches. Furthermore, flow reversal and secondary motions have been detected throughout the cardiac cycle, making this circulatory configuration far from intuitive [9]. Surgeons are therefore deeply interested in understanding the local hemodynamics of the Hybrid Norwood anatomy and the potential benefits of the reverse BT shunt, especially when severe stenosis is present. The main objectives of this study are therefore: 1) To develop a closed-loop, multiscale model of the cardiovascular system able to describe detailed local hemodynamics and its effects on

Figure 1: HLHS Anatomy (left)[4], Standard Norwood Anatomy (center)[5], Hybrid Norwood Anatomy (right)[6]

Figure 2: Hybrid Norwood Anatomy with RBTS in place
the global circulation; 2) assess the performance of the RBTS in increasing perfusion to the coronary and carotid circulation.

**Materials and Methods**

**Anatomical Model**

The aortic arch anatomy representative of an infant after undergoing the Hybrid Norwood procedure was constructed using SolidWorks (Dassault Systemes, Concord, MA) and includes the pulmonary root (PR), descending aorta (DA), innominate artery (IA), right and left subclavian arteries (RSA, LSA), right and left carotid arteries (RCA, LCA), and right and left coronary arteries (RcorA, LcorA) (Figure 3). Three other models were constructed, one with severe stenosis, one with the RBTS, and one with severe stenosis and RBTS (Figure 3).

**Computational Fluid Dynamics Model**

The solid models were then imported into Star-CCM+ (CD-adapco, New York, NY), a CFD software that uses the finite volume method to discretize the solution domain. The models were meshed using a hexahedral mesh with fine prism-layer elements near the walls (modeled as rigid) for better boundary layer resolution. The number of elements for the different meshed geometries varied in the range from 1 to 1.2 million. A fully viscous Realizable K-epsilon turbulence model with enhanced wall treatment was adopted. Blood was modeled as a Newtonian and incompressible with a density of $1060 \text{ kg/m}^3$ and a viscosity of 0.004 Pa-s. An unsteady, implicit Navier-Stokes equation solver was used over several cardiac cycles, the duration of which is 0.462 seconds. The time step used to advance the solution was $4.62 \times 10^{-3}$. Unsteady stagnation pressure inlet for the cardiac output and unsteady flow splits were utilized as boundary conditions derived from the lumped parameter model described in the following section.

**Lumped Parameter Model**

The circulation of the entire body was modeled using Greenfield-Fry’s electrical analogy, where the viscous drag is modeled as a resistor, flow inertia as an inductor, vessel compliance as a capacitor, and heart valves as diodes (Figure 4). The ventricle is modeled as a time varying compliance called elastance; this provides the pulsatile nature of the cardiac output and is the driving function of the circuit. The different vascular beds are modeled as RLC compartments (Figure 5) with the whole circuit tied together in a closed loop representation of physical anatomy. The resulting circuit contains 32 state variables and the ensuing system of ordinary differential equations is solved using a 4th Order Runge-Kutta integrator with adaptive time step. The coupling of the two models is done in the following manner: 1) The initial circuit is tuned to produce target flows and pressure waveforms obtained from catheter studies and surgeon’s criteria, 2) Flow splits and inlet boundary conditions are imposed to the CFD from the circuit, 3) CFD simulation is carried out to obtain pressure waveforms, 4) The CFD equivalent parameters within the circuit are modified to match those derived from the CFD solution, 5) Impose new flow splits from circuit to CFD, 6) Iterate until convergence. Once this iterative process is complete, the CFD simulation is run for 3 cardiac cycles and post-processing is performed.
Results

The flow and pressure waveforms obtained for the different configurations at the boundaries of the CFD model are shown in Figure 6. A summary of cardiac output, the relative flow rates through the shunt, branching arteries, and the ratio of pulmonary to systemic circulation is presented in Table 1. Table 2 quantifies the percent change in flow relative to the nominal case. As expected, there is a significant reduction in both pressure and net flow rate through the arteries of interest when the severe stenosis is present. The performance of the RBTS shunt is highly satisfactory, returning the pressure and flow rate values to near stenosis-free levels. Notice there are no adverse effects when implementing the RBTS without stenosis, on the contrary, there is a slight increase in flow rate to both arterial beds. This supports the idea of implanting the shunt as a preventative measure in case stenosis develops with time after the initial procedure, reducing or eliminating the need for re-intervention. Of concern to surgeons was the possible “siphoning” effect that might occur during diastole, in which the lower pressure in the pulmonary circulation creates flow reversal and could be exacerbated by the addition of the RBTS [9]. As seen in Figure 6, there is a slightly higher amount of reverse flow with the RBTS in place with the Nominal configuration but this is offset by higher peak flow. The same behavior is not exhibited in the other configurations.

![Figure 6: Pressure and flow rate waveforms at carotid and coronary artery boundaries](image)

Table 1: Flow rate summary for all anatomical configurations

<table>
<thead>
<tr>
<th>Cardiac Output (ml/cycle)</th>
<th>Flow Rate as Percentage of Cardiac Output</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>DA</td>
</tr>
<tr>
<td>Nominal</td>
<td>2005</td>
</tr>
<tr>
<td>Nominal BT-Shunt</td>
<td>2025</td>
</tr>
<tr>
<td>Severe</td>
<td>1902</td>
</tr>
<tr>
<td>Severe BT-Shunt</td>
<td>2001</td>
</tr>
</tbody>
</table>
The CFD study provides detailed visualization capabilities of the unsteady flow field. Of particular interest to surgeons are the areas of flow stagnation, impingement and high/low shear stress. Studies [10] have shown a strong correlation in the magnitude of shear stress, endothelial cell function, and vessel wall remodeling. Low levels of shear stress promote platelet activation and plaque buildup and coincide with areas of low flow velocity, typically the outer walls of bifurcations and recirculation zones. Varying levels of shear stress through the ductus arteriosus, where a metallic stent is placed to prevent it from receding, affect the formation of neointimal hyperplasia and stenosis. Flow characteristics surrounding the RBTS anastomosis are also of great interest since there has been early evidence of pulmonary root hyperplasia and studies have suggested there is an optimal shunt diameter as a function of shear stress that achieves greater shunt patency and reduction in graft thrombosis [11]. The following figures help identify such areas of interest: Figure 8 is a streamline plot with seed point at the pulmonary root and Figure 9 is a wall shear stress contour plot. Figure 7 depicts the points during the cardiac cycle where the aforementioned plots were taken. Point 1 occurs at peak systole, 2 during diastole where ventricular isovolumic relaxation occurs, 3 during diastole when a small amount of retrograde flow is seen across the pulmonary valve as it closes, and 4 at the onset of ventricular systole.

Table 2: Flow rate difference with respect to nominal configuration

<table>
<thead>
<tr>
<th></th>
<th>Cardiac Output</th>
<th>DA</th>
<th>LCA</th>
<th>LcorA</th>
<th>LPA</th>
<th>LSA</th>
<th>RCA</th>
<th>RcorA</th>
<th>RPA</th>
<th>RSA</th>
<th>Qs/Qp</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nominal BT-Shunt</td>
<td>0.97</td>
<td>0.77</td>
<td>1.50</td>
<td>1.76</td>
<td>0.73</td>
<td>1.50</td>
<td>1.77</td>
<td>0.76</td>
<td>1.76</td>
<td>0.73</td>
<td>0.41</td>
</tr>
<tr>
<td>Severe</td>
<td>-5.15</td>
<td>2.26</td>
<td>-29.35</td>
<td>-30.45</td>
<td>1.39</td>
<td>-28.37</td>
<td>29.28</td>
<td>-30.45</td>
<td>1.39</td>
<td>-29.30</td>
<td>-12.48</td>
</tr>
<tr>
<td>Severe BT-Shunt</td>
<td>-0.21</td>
<td>0.06</td>
<td>-0.93</td>
<td>-0.24</td>
<td>-0.22</td>
<td>-0.98</td>
<td>-0.30</td>
<td>-0.24</td>
<td>-0.22</td>
<td>-0.32</td>
<td>0.02</td>
</tr>
</tbody>
</table>

Figure 7: Cardiac output waveform

Figure 8: Streamline plot with seed point at the pulmonary root

Figure 9: Wall shear stress contour plot
Figure 8: Streamlines colored by velocity magnitude

Figure 8: Wall shear stress magnitude contour plots
Conclusions and Future Work

The multiscale model developed in this study shows its tremendous potential as a predictive tool for surgeons. Detailed local hemodynamics derived from the model provide an insight of the effects of severe stenosis and RBTS implantation (Figure 7). Though the present analysis concerned changes in the local anatomy, the model can also be used to assess changes in controllable physiological responses affecting peripheral vascular beds, such as cerebral vascular resistance. Patient specific anatomy can also be used instead of the synthetic models to provide analysis on an individual basis. Figure 8 shows a 3D models from CT-Scans performed by this research team of the anatomy of an infant after the Hybrid Norwood procedure. The evolution of the current model is to implement patient specific anatomy with Fluid Structure Interaction algorithms that enable compliance of the walls to be taken into account.

References


