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Running head: Model of Hybrid Norwood Circulation

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Abstract

Background:

Hybrid Norwood (HN) procedure hemodynamic characteristics differ from the conventional Norwood and are not fully understood. We present a multi-scale model of HN circulation to understand the local hemodynamics and effects of aortic arch stenosis and a reverse-BT shunt (RBTS) on coronary and carotid perfusion.

Methods:

Four 3D models of four HN anatomic variants were developed— with/without 90% distal preductal arch stenosis and with/without a 4mm RBTS. A lumped parameter model (LPM) of the circulation was coupled to a local 3D computational fluid dynamics (CFD) model. The outputs from the LPM provided waveform boundary conditions for the CFD model.

Results:

A 90% distal arch stenosis reduced pressure and net flow-rate through the coronary and carotid arteries by 30%. Addition of the RBTS completely restored pressure and flow rate to baseline in these vessels. Zones of flow stagnation, flow reversal, and recirculation in the presence of stenosis were rendered more orderly by addition of the RBTS. In the absence of stenosis, presence of the shunt resulted in extensive zones of disturbed flow within the RBTS and arch.

Conclusions:

We found that a 4mm x 21mm RBTS completely compensated for the effects of a 90% discrete stenosis of the distal aortic arch in the HN. Placed "preventatively", the RBTS and arch displayed zones with thrombogenic potential showing recirculation and stagnation that persist for a substantial fraction of the cardiac cycle.

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230 words.

Introduction

The hybrid Norwood (HN) management of hypoplastic left heart syndrome (HLHS) has emerged as a promising strategy. The procedure avoids cardiopulmonary bypass and consists of branch pulmonary artery banding, stenting of the ductus arteriosus and balloon atrial septostomy [1]. The less invasive nature of the HN and deferment of the risk of major open heart surgery to an older age are considered to help improve survival, neurological and cardiac functional outcomes [2]. Most recent reports of survival results after the HN are comparable with those obtained with the traditional surgical Norwood and, at some institutions HN has become the preferred first intervention for HLHS palliation [3-5].

An important concern after the HN is the possibility of obstruction in the aortic isthmus after stent deployment. This is because during the inter-stage period the coronary and cerebral circulations depend mainly on retrograde flow through the aortic arch. In patients with aortic atresia this flow is entirely dependent on retrograde perfusion. Obstruction can occur immediately as a result of stent maldeployment, within a few hours from ductal remodeling after discontinuation of prostaglandins, or late as a result of fibrosis in the distal stent [6]. Clinically important obstruction of the distal aortic arch has been reported to occur in 24% of patients after hybrid procedures for HLHS [7]. It has been suggested that, in patients that develop distal aortic arch obstruction, placement of a reverse Blalock-Taussig shunt (RBTS, main pulmonary artery-to-innominate artery shunt) could prevent myocardial and cerebral ischemia [6]. The RBTS is a straightforward surgical addition to the HN and, although the benefits of implementing this step remain unproven, some groups have adopted the policy of placing this shunt as a prophylactic measure in patients with limited or absent antegrade aortic flow, or at high risk of developing aortic arch obstruction [2, 6].

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Computational fluid dynamics (CFD) is being successfully employed to elucidate the optimal approach to staged reconstruction of HLHS [8-12]. We extend these investigations by employing a multi-scale CFD model of the HN to characterize the effects of stenosis of the distal arch and of the RBTS on cerebral and coronary perfusion as well as the local flow patterns through the aortic arch and the RBTS graft.

Material and Methods

Complex hemodynamics following HN are examined via a multi-scale model: lumped parameter model (LPM) of the peripheral circulation coupled with a localized 3D-CFD model.

Anatomical Model

Synthetic rigid-walled 3D models representative an infant with HLHS following the HN procedure were constructed using SolidWorks (Dassault Systemes, Concord, MA) assuming atresia of the aortic valve and including the ascending aorta (AA), transverse arch (TA), innominate artery (IA), right and left subclavian arteries (RSA, LSA), right and left carotid arteries (RCA, LCA), main pulmonary artery (MPA), branched pulmonary arteries (BPA, right=RPA, left=LPA), patent ductus arteriosus (PDA), descending aorta (DA), and right and left coronary arteries (RcorA, LcorA).

Four models were developed (Figure 1A):

The nominal model (Nom) corresponds to the standard HN configuration of banded BPA and stenting of the PDA with "typical" hypoplasia of the aortic arch. In the second model (Sten), part of the computational domain was removed at a point proximal to the PDA and distal to the LSA (aortic isthmus) to represent severe discrete stenosis. A reduction in lumen cross-sectional area of 89.3% was produced through an aortic isthmus reduction from a nominal area of 21.46 mm^2 to 2.30mm^2 (Figure 1B). The third and fourth models (Nom-RBTS, Sten-RBTS) were developed

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by modifying the Nom and Sten models respectively incorporating a RBTS (4mm x 21mm bypass graft from the MPA to the IA). The proximal end of the RBTS was modeled proximal to the pulmonary bifurcation and anastomosed to the anterior wall of the MPA. The distal graft was anastomosed to the IA in an end-to-side fashion. A BPA band diameter of 2mm was used. Vessel diameters and other important dimensions along the aortic arch are depicted in Figures 2A and 2B.

CFD Model

Solid models were imported into Star-CCM+ (CD-Adapco, NY), a commercial Finite Volume-based CFD software. A high quality mesh was obtained for all models providing grid-independence and adequate capture of the boundary layer and detailed flow features [13, 14]. Figure 2C shows overall element distribution and mesh detail.

Blood was modeled as an incompressible Newtonian fluid with density of $\rho=1060kg/m^3$ and viscosity of $\mu=0.004Pa\text{-}s$. The 3D flow field is obtained by numerically resolving the Navier-Stokes (NS) mass and momentum conservation equations:

$$(1)$$
$$(2)$$

Here, is the velocity vector and is the pressure field. The NS were solved with an unsteady implicit scheme. The time step of 4.62ms provided time-independent solution for a 130 bpm. Waveforms provided by the LPM are used to impose an unsteady stagnation pressure inlet at the MPA root and prescribe unsteady flow splits as arterial outlet boundary conditions.

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Lumped Parameter Model

The LPM is an electrical analog of the circulatory system [15] modeling viscous drag as a resistor (R), flow inertia as an inductor (*L*), vessel compliance as a capacitor (*C*), and tricuspid and pulmonary valves as ideal diodes (Figure 3A). A pair of differential equations governs each *R-L-C* compartment model of a vascular bed:

$$\text{(3)}$$

where, *Q* is the flow-rate and *Δp* is the pressure difference, while the second equation models vessel wall compliance with $C = \dfrac{dV}{dp}$. Previously published work by others provided baseline values of *R*, *L*, and *C* [8-12]. These were adjusted iteratively to approach waveforms from catheterization data of a "typical" HN patient. The arterial vascular bed resistance is tuned first, since it is the primary determinant of total flow through any given artery. Compliance and inertance parameters are then tuned to approach the desired waveform. Comparable methods have been successfully utilized in previous CFD studies of palliative strategies in HLHS [8-12].

The right ventricle, modeled as a time varying capacitor, , is the driving function of the circuit providing pulsatile cardiac output. Its reciprocal, the elastance, , relates ventricular pressure and volume at a given point during the cardiac cycle [16]. We used the form , where, is the "double hill" normalized elastance function which has been modified from the adult model in [17] to a neonate model as:

$$\text{(4)}$$

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Where, , , and *HR* is the heart rate , , and the exponential coefficients in Eq. (4) were iteratively tuned to produce a cardiac output for the nominal model of 2.0 These values were held constant for all subsequent simulations.

A 32 state variable closed-loop circuit representation of the systemic and pulmonary circulation (Figure 3A) leads to coupled ordinary differential equations that are solved via a 4th order adaptive Runge-Kutta integrator. BPA banding is achieved using a geometrical restriction in the CFD model supplemented with a resistance placed in pulmonary LPM vascular bed to achieve cycle-averaged ratio of total BPA artery to ductal flow, Qp/Qs~1, in the nominal model. In order for the LPM to account for most of the coronary perfusion during diastole, the coronary arterial bed resistance is assumed to be a normalized exponential function of the time-varying elastance. Figure 4 compares ventricular and aortic pressure waveforms produced by the LPM and the corresponding waveforms from a typical HN patient. The nominal LPM parameters are held constant in subsequent simulations in which the RBTS, as well as various levels of stenosis in the isthmus, are introduced into the 3D model.

Coupling

Coupling is achieved by: (1) tuning the initial circuit to produce target flows and pressure waveforms obtained from catheter data when available supplemented with nominal values for typical HLHS patients, (2) imposing transient flow splits and inlet boundary conditions to the CFD model from the circuit, (3) carrying out the CFD simulation to obtain pressure waveforms, (4) modifying the CFD equivalent parameters within the circuit to match those derived from the CFD, (5) imposing new flow splits from circuit to CFD, and (6) iterating the system of equations until convergence (Figure 3B). Convergence is achieved once the relative change in flow rates

at all branch vessels is less than 10^{-2}, reached typically in 20-30 iterations between the CFD and LPM models. Once the process has converged, the CFD simulation is run for three cardiac cycles to achieve a sustained periodic solution, and post-processing is performed. Our iterative approach provides a convenient and computationally effective way to tune the LPM parameters.

Results

Calculated flow and pressure waveforms are presented for all models in Figures 5-6, and Table 1. Values for the Nom model resulting from iteratively tuning the lumped and CFD models provide representative HN conditions, namely Qp/Qs~1 and a CO~2.0 liters/min as well representative arterial outlet waveforms.

The set of composite plots in Figure 5A are the driving waveforms of the computational model. Figure 5B presents a composite pressure plot of the pulmonary root and all arterial outlets of the four CFD models. The LPA and RPA pressures distal to the bands are calculated by using the pressures in Figure 5B and reducing them by the gradient imposed in the pulmonary compartment to model the bands themselves. The resulting peak systolic pressures ranged from 12-15mmHg.

The instantaneous flow rates and corresponding pressure traces over a single cycle for each vessel and each model are displayed in Figure 6A-F. (The flow rates for the right and left-sided arch vessels were averaged for these plots). The pulsatility in all of the arch branches is significantly blunted in the Sten model but is restored in the presence of the RBTS.

Individual cycle-averaged flow rates are provided in Table 1 as a percentage of cardiac output as well percentage of change from those of Nom.

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Although there is an important reduction in both pressure and net flow rate through the coronaries and carotids when severe stenosis is present, addition of the RBTS returns the pressure, flow rate and pulsatility nearly to Nom values. Blood flow and pressure profiles were not substantially changed when incorporating the RBTS in the absence of stenosis. In particular, the RBTS did not "siphon" net blood flow from the arch branches or coronaries.

Figures 7-9 show streamlines superimposed with velocity vectors for the four models at selected points during the cardiac cycle: 1) peak systole, 2) early diastole, 3) mid-diastole and 4) late diastole.

The streamline plots of Nom in Figure 7A reveal, in peak systole, flow acceleration in the TA proximal to the origin of the LCA where an impingement zone is observed. An area of low flow posterior and just distal to the LSA persists throughout most of the cardiac cycle. Curiously, this corresponds to the area most subject to recurrent stenosis in the HN.

The flow field for Nom-RBTS is displayed in Figure 7B. Here, prominent recirculation and stagnation zones are observed, particularly at the origin of IA caused by the confluence of retrograde flow from the PDA and the RBTS through the IA, as well as at the distal anastomosis of the RBTS. Both these recirculation patterns remain present through most of the cardiac cycle. Moreover, the flow in the RBTS is seen to feature chaotic swirl in early diastole which progressively increases in mid to late diastole, at which point the flow stagnates and reverses in direction. These phenomena can be appreciated in companion Figure 9A.

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In the case of Sten model seen Figure 8A, a high velocity jet emanates retrograde with a peak velocity of near 4m/s during peak systole impinging proximal to the root of the LSA. Several recirculation zones are seen in the aortic arch and proximal portions of the arch branches that persist for much of the cardiac cycle. These zones are not present in the absence of distal arch stenosis.

The flow field for the Sten-RBTS is displayed in Figure 8B. Here, retrograde flow from the PDA jets through the stenosis and is mainly ingested by the LSA. Flow from RBTS is distributed to the innominate artery as well as the aortic arch. The arch flow splits between the coronaries, and the LCA where it confluxes with the retrograde flow from the PDA. It is noted that the recirculation features seen in the Nom-RBTS model are no longer present in the Sten-RBTS model. Only a discrete area of stagnation and impingement is still noted at the root of the LSA. A close up of the flow field in the RBTS itself is displayed in Figure 9B. The chaotic swirling seen through the shunt in the Nom-RBTS model is not present in the Sten-RBTS. This is a consequence of the flow velocity in the RBTS in both cases. The maximum velocity seen in the Nom-RBTS case through the shunt in peak systole is around 1.3 m/s while in the Sten-RBTS the flow through the shunt has a higher velocity in peak systole of about 2.7 m/s and maintains a maximum of 1.2 m/s in mid-diastole.

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Discussion

The HN scheme for staged palliation of HLHS, pioneered by Galantowicz and others [1-7, 18], obviates the need for major cardiovascular surgery in the newborn period. Although the standard and hybrid Norwood basic objectives are the same, the resulting anatomical configurations are clearly different. This raises the possibility of distinct differences in the resulting hemodynamics and physiology and unique effects on the various arterial segments of the reconstruction. Additionally, the HN is subject to unique complications such as distal arch obstruction, ductal stenosis, stent migration, atrial septal restriction, and under- or over-circulation to one or both lungs, all of which are known to have a substantial impact on early and intermediate term morbidity and mortality. Thus, it is important to understand the hemodynamic properties of the HN.

Whereas clinical imaging and catheter measurements provide some understanding, CFD has the power to elucidate hemodynamic behavior in a controlled manner and at sub-millimeter level of detail [8-12]. Corsini et al [18] examined the effects of various degrees of branch pulmonary artery banding and ductal stent diameter on cardiac output and oxygen delivery in the HN model. They found that these parameters were much more sensitive to changes in the percentage constriction of the BPA than to the same percentage change in ductal diameter, and suggested that a banded BPA lumen diameter of 2mm was optimal. Subsequently Hsia et al [19] compared cardiac output, systemic and cerebral oxygen delivery in two variants of the standard Norwood and the HN circulations. Controlling for all other characteristics of cardiac function and peripheral beds, they found that cardiac output and oxygen delivery were significantly lower in the HN circulation, despite the presence of an unobstructed aortic arch. These findings are consistent with the clinical reports [20-21]. This illustrates the power of CFD

to demonstrate underlying mechanisms, not necessarily intuitive, for the clinical findings in this case that the combination of diastolic flow reversal in the ductus, plus obligatory retrograde (albeit unobstructed) arch flow was responsible for reductions in both cardiac output and cerebral blood flow.

Using similar HN models and CFD, the present work adds to the findings of others [12, 18]. In our HN study, we considered the effects of both distal aortic arch obstruction and the RBTS on the hemodynamics of the aortic arch and its branches. Distal aortic arch obstruction occurs in 15-25% [5, 7, 19] of patients who have undergone the HN procedure. Causes may include juxtaductal intimal or ductal cell proliferation, stent malposition, or chronic flow disturbance. Caldarone et al [6] first proposed treating, or even preventing the effects of distal arch obstruction by suturing a graft (RBTS) from the pulmonary trunk or bifurcation area to the IA.

In the present work, we modeled distal arch obstruction as a discrete 90% reduction in luminal cross sectional area. The subsequent CFD calculations showed that this resulted in an approximately 30% reduction in mean flow and pressure in the coronary arteries and all arch branches, and a 5% reduction in cardiac output. Models including arch stenosis have lower cardiac output due to the increased afterload resulting from the reduction of the isthmus lumen. The presence of a 4mm x 21mm RBTS completely compensated for the distal arch obstruction, restoring cardiac output, coronary and arch branch flow and pressure to nominal levels. Additionally, we examined the flow fields in detail and found that, whereas distal arch obstruction produced several zones of recirculation and stagnation (each several mm in size) in the aortic arch (as opposed to the Nom model), the RBTS eliminated these zones, rendering the flow more orderly. The RBTS, therefore, may serve the additional advantage of eliminating abnormal flow zones that, in the setting of distal arch obstruction, may promote thrombotic arch

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occlusion or thromboembolism.

We constructed the Nom-RBTS model to represent the situation where a RBTS is placed "preventatively", before significant distal arch obstruction develops. In this case the RBTS did not change the overall hemodynamics substantially. In particular, whereas the position of the RBTS did slightly increase diastolic flow reversal in the ascending aorta (potentially reducing coronary perfusion), this seemed to be offset by greater peak antegrade flow.

On the other hand, in the detailed examination of the flow field in the Nom-RBTS model, we found that the RBTS resulted in recirculation zones in the IA and part of the aortic arch (Figure 7B). Additionally, the flow within the graft was quite chaotic with several recirculation zones and of low velocity for a good fraction of the cardiac cycle (Figure 9A). Our prior calculations showed that such zones were characteristically those of low shear stress promoting platelet activation, aggregation and thrombosis, especially within the lumen of a synthetic graft [23, 24]. Therefore, our findings suggest that a RBTS placed "preventatively" may be at increased risk for thrombosis and occlusion. In the HN circulation, such thrombi may embolize to the brain due to the prevailing direction of flow in the graft.

Our study has several limitations. First we did not study the effect of various degrees of stenosis, or of different RBTS graft diameters, lengths and anastomotic configurations. A future extended analysis of this type might reveal an "optimal" configuration of the RBTS to minimize its chance of thrombotic occlusion. Second, our rigid-walled CFD model did not account for arterial elasticity (although our LPM model provides an inlet boundary condition that accounts for MPA compliance). Vessel wall elasticity can exaggerate impedance mismatches at vessel junctions or anastomoses, thereby affecting the flow characteristics. We intend to include the

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effects of vessel-wall compliance in the CFD model through the addition of a fluid structure interaction algorithm in our future work. Also, we are considering performing an uncertainty and sensitivity analysis to determine the robustness of the model to the fitted data.

Conclusions

Using CFD, we have studied the HN circulation emphasizing the effects of distal aortic arch obstruction and/or the presence of a RBTS. We found that a 4mm x 21mm RBTS completely compensates for the effects of a 90% discrete stenosis of the distal aortic arch. Placed "preventatively", however, the RBTS and arch displayed recirculation and stagnation zones persisting for a substantial fraction of the cardiac cycle. Such zones have been suggested to promote thrombus formation.

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Acknowledgements

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Disclosures

None.

References

1. Galantowicz M. The hybrid approach to hypoplastic left heart syndrome. Operative techniques in thoracic and cardiovascular surgery. 2009;74-85.

2. Alsoufi B, Bennetts J, Verma S, Caldarone CA. New developments in the treatment of hypoplastic left heart syndrome. Pediatrics. 2007;119(1):109-17.

3. Barron DJ, Kilby MD, Davies B, Wright JG, Jones TJ, Brawn WJ. Hypoplastic left heart syndrome. Lancet. 2009;374(9689):551-64.

4. Caldarone CA, Benson L, Holtby H, Li J, Redington AN, Van Arsdell GS. Initial experience with hybrid palliation for neonates with single-ventricle physiology. Ann Thorac Surg. 2007;84(4):1294-300.

5. Galantowicz M, Cheatham JP, Phillips A, et al. Hybrid approach for hypoplastic left heart syndrome: intermediate results after the learning curve. Ann Thorac Surg. 2008;85(6):2063-70.

6. Caldarone CA, Benson LN, Holtby H, Van Arsdell GS. Main pulmonary artery to innominate artery shunt during hybrid palliation of hypoplastic left heart syndrome. J Thorac Cardiovasc Surg. 2005;130(4):e1-2.

7. Stoica SC, Philips AB, Egan M, et al. The retrograde aortic arch in the hybrid approach to hypoplastic left heart syndrome. Ann Thorac Surg. 2009;88(6):1939-46.

8. Migliavacca F, Pennati G, Dubini G, et al. Modeling of the Norwood circulation: effects of shunt size, vascular resistances, and heart rate. Am J Physiol Heart Circ Physiol. 2001;280(5):H2076-86.

9. Migliavacca F, Laganà K, Pennati G, de Leval MR, Bove EL, Dubini G. Global mathematical modelling of the Norwood circulation: a multiscale approach for the study of the pulmonary and coronary arterial perfusions. Cardiol Young. 2004;14 Suppl 3:71-6.

10. Laganà K, Balossino R, Migliavacca F, et al. Multiscale modeling of the cardiovascular system: application to the study of pulmonary and coronary perfusions in the univentricular circulation. J Biomech. 2005;38(5):1129-41.

11. Bove EL, Migliavacca F, de Leval MR, et al. Use of mathematic modeling to compare and predict hemodynamic effects of the modified Blalock-Taussig and right ventricle-pulmonary artery shunts for hypoplastic left heart syndrome. J Thorac Cardiovasc Surg. 2008;136(2):312-320.e2.

12 Hsia TY, Cosentino D, Corsini C, Pennati G, Dubini G, Migliavacca F. Use of mathematical modeling to compare and predict hemodynamic effects between hybrid and surgical Norwood palliations for hypoplastic left heart syndrome. Circulation. 2011;124(11 Suppl):S204-10.

13. Osorio AF, Osorio R, Ceballos A, et al. Computational fluid dynamics analysis of surgical adjustment of left ventricular assist device implantation to minimize stroke risk. Comput Methods Biomech Biomed Engin. 2011 *In press*.

14 Chung, TJ. Finite Volume Methods. In Computational Fluid Dynamics. 2nd ed. Cambridge: Cambridge UP; 2002: 227-29.

15. Waite L, Fine J. Pulsatile flow in Large Arteries. In Applied Biofluid Mechanics. 1st ed. New York, MY: McGraw-Hill;2007:187-221.

16. Suga H, Sagawa K. Instantaneous pressure-volume relationships and their ratio in the excised, supported canine left ventricle. Circulation Research. 1974;35(1):117-26.

17. Simaan MA, Ferreira A, Chen S, Antaki JF, Galati DG. A dynamical state space representation and performance analysis of a feedback-controlled rotary left ventricular assist device. IEEE Transactions on Control Systems Technology. 2009;17(1):15-28.

18. Corsini C, Cosentino D, Pennati D. Multiscale models of the hybrid palliation for hypoplastic left heart syndrome. J Biomech. 2011;44:767-770.

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19. Bacha EA, Daves S, Hardin J, et al. Single-ventricle palliation for high-risk neonates: the emergence of an alternative hybrid stage I strategy. J Thorac Cardiovasc Surg. 2006;131(1):163-171.e2.

20. Li J, Zhang G, McCrindle B. Profiles of hemodynamics and oxygen transport derived by using continuous measured oxygen consumption after the Norwood procedure. J Thorac Cardiovasc Surg. 2007;133(2):441-8.

21. Honjo O, Caldarone CA. Hybrid palliation for neonates with hypoplastic left heart syndrome: current strategies and outcomes. Korean Circ J. 2010;40(3):103-11.

22. Egan MJ, Hill SL, Boettner BL, et al. Predictors of retrograde aortic arch obstruction after hybrid palliation of hypoplastic left heart syndrome. Pediatr Cardiol. 2011;32(1):67-75.

23. Malek AM, Alper SL, Izumo S. Hemodynamic shear stress and its role in atherosclerosis. JAMA. 1999;282(21):2035-42.

24. Binns RL, Ku DN, Stewart MT, Ansley JP, Coyle KA. Optimal graft diameter: effect of wall shear stress on vascular healing. J Vasc Surg. 1989;10(3):326-37.

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Tables

Table 1: Cardiac output, flow distribution and flow changes

			Flow rate as percentage of cardiac output									
Model	Cardiac output (ml/min)	Qp/Qs	DA	LCA	LcorA	LPA	LSA	RCA	RcorA	RPA	RSA	Shunt
Nominal	2015	0.94	29.4	4.6	1.8	24.1	4.7	4.6	1.8	24.3	4.7	
Nom-RBTS	2022	0.93	29.4	4.7	1.8	24.0	4.7	4.6	1.8	24.3	4.7	11.6
Stenosis	1863	1.10	31.9	3.2	1.3	26.1	3.3	3.2	1.3	26.3	3.2	
Sten-RBTS	1975	0.94	29.7	4.5	1.8	24.1	4.6	4.6	1.8	24.3	4.6	21.1

Percentage of flow change from Nominal

Nom-RBTS	0.35	-0.2	0.3	0.5	0.3	0.3	0.5	0.7	0.3	0.3	0.7	
Stenosis	-7.54	17.8	0.4	-35.7	-31.7	0.3	-35.7	-35.7	-31.7	0.3	-35.7	
Sten-RBTS	-1.99	0.3	-1.0	-4.3	-2.1	-1.8	-4.4	-3.7	-2.0	-1.8	-3.7	

Average Blood Flow Velocity Through Shunt (m/s)

	Cardiac cycle	Systole to Early Diastole (Antegrade flow, PA to IA)	Mid to Late Diastole (Retrograde flow, IA to PA)
Nom-RBTS	0.26	0.47	0.10
Sten-RBTS	0.62	1.32	0.27

DA=descending aorta, LCA=left carotid, LcorA=left coronary, LPA=left pulmonary, LSA=left subclavian, Qp/Qs=pulmonary to

systemic flow ratio, RCA=right carotid, RcorA=right coronary, RPA=right pulmonary, RSA=right subclavian.

Figure Legends

Figure 1: (A) 3D models of the HN circulation. (B) Detailed view of the transverse aortic arch with stenosis highlighted in red. DA=descending aorta, LCA=left carotid artery, LcorA=left coronary artery, LPA=left pulmonary artery, LSA=left subclavian artery, RCA=right carotid artery, RcorA=right coronary artery, RSA=right subclavian artery, RPA=right pulmonary artery.

Figure 2: Dimensions: (A) Sten and (B) Nom-RBTS models. (C) Sten-RBTS model computational domain mesh. Measurements in millimeters. Ø=diameter.

Figure 3: Multi-scale model of the HN. (A) 3D model coupled with LPM. (B) coupling scheme.

Figure 4: Ventricular and descending aorta pressures produced by the multi-scale model (A and C) and from typical HN patient (B and D).

Figure 5: Composite plots: (A) ventricular, atrial, and pulmonary root pressure (left scale) with cardiac output (right scale). (B) Pressure traces of all arteries. DA=descending aorta, LCA=left carotid artery, LcorA=left coronary artery, LPA=left pulmonary artery, LSA=left subclavian artery, P.Root=pulmonary root, RCA=right carotid artery, RcorA=right coronary artery, RSA=right subclavian artery, RPA=right pulmonary artery.

Figure 6: Flow-rates and pressures over a cardiac cycle: (A) coronaries, (B) carotids, (C) subclavians, (D) main pulmonary, (E) descending aorta, and (F) Reverse-BT shunt.

Figure 7: Streamlines at selected times in the cardiac cycle: (A) Nom, (B) Nom-RBTS models.

Figure 8: Streamlines at selected times in the cardiac cycle: (A) Sten, (B) Sten-RBTS.

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Figure 9: Streamline details at selected times in the cardiac cycle: (A) Nom-RBTS, (B) Sten-RBTS.

Abbreviations

AA	Ascending aorta
BPA	Branched pulmonary arteries
CFD	Computational Fluid Dynamics
DA	Descending aorta
HLHS	Hypoplastic left heart syndrome
HN	Hybrid Norwood
IA	Innominate artery
LCA	Left carotid artery
LcorA	Left coronary artery
LPA	Left pulmonary artery
LPM	Lumped Parameter Model
LSA	Left subclavian artery
MPA	Main pulmonary artery
Nom	Nominal anatomy
Nom-RBTS	Nominal with RBTS anatomy
NS	Navier-Stokes equations
PDA	Patent ductus arteriosus
RBTS	Reverse-BT shunt
RCA	Right carotid artery
RcorA	Right coronary artery
RPA	Right pulmonary artery
RSA	Right subclavian artery
Sten	Stenosis anatomy
Sten-RBTS	Stenosis with RBTS anatomy
TA	Transverse arch