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Numerical Simulation and Hemodynamic Analysis of the Modified Blalock-Taussig Shunt

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Abstract—**The modified Blalock-Taussig (mB-T) shunt is an effective palliative surgical method in the treatment of cyanotic congenital heart diseases. It increases the pulmonary blood flow through an implanted shunt between systemic and pulmonary arteries. The surgical technique improved over the years. However, it is still a challenge to control appropriate distribution of blood flow through this shunt after this kind of procedure till now. Here, we report on the method of computational fluid dynamics (CFD) for the hemodynamic studies of a patient-specific case after the mB-T shunt. The analysis system that we validated previously in the studies of the Norwood procedure was applied to predict the hemodynamic characteristics in the mB-T hunt area. The real-time velocities derived from Echocardiography measurements and the blood pressure wave reflections from peripheral vessels were utilized as boundary conditions to physiologically capture the blood flow information in the simulation. The local pressure, blood flow distribution and wall shear stress were calculated. The results suggest pressure decreases greatly through the shunt and around 40% of blood flow is distributed from the systemic circulation to pulmonary arteries in one cardiac cycle. These indict computational hemodynamics may be applied in future studies of establishing quantitative standards to evaluate the outcomes of the mB-T shunt and to optimize the implantation of the mB-T shunt in virtual surgeries.**

Keywords—**Modified Blalock-Taussig shunt, computational fluid dynamics, hemodynamics, cardiovascular flow**

I. Introduction

The modified Blalock-Taussig (mB-T) shunt is an effective surgical method for the patients with cyanotic congenital heart diseases. It is a palliative procedure to increase pulmonary blood flow and alleviate cyanosis in surgical planning for severe cyanotic congenital heart diseases, usually serving as the first step of staged procedures. By surgically connected the systemic and pulmonary circulations through a Gore-Tex conduit, blood from the systemic arteries can flow into the lungs to improve the oxygen saturation of the arterial blood.

After Blalock and Taussig [1] introduced the classical B-T shunt between the subclavian artery and the pulmonary artery in 1945, various modifications aimed to construct a systemic pulmonary shunt have been described and applied, such as direct aortopulmonary anastomoses (Waterston and Potts shunts), interposition prosthetic conduits. However, a modified form of the B-T shunt is the preferred procedure in clinical applications [2, 3] over the past 70 years. In the mB-T shunt, the end-to-side anastomosis was made between a Gore-Tex conduit and the subclavian artery or the innominate artery and the pulmonary artery respectively, without cutting off the subclavian artery. Due to the fixed diameter of the conduit, the distribution of blood flow through this conduit is easy to be controlled by choosing appropriate conduit, thus in voiding excessive or insufficient blood flow distribution to the lungs.

Although survival rate after the mB-T shunt has been improved over the years, a number of unresolved questions remain regarding optimal shunt size, shunt material, distortion of the pulmonary arteries, potential risk of arrhythmia, and so on. These questions were partly due to limited quantitative analysis of the hemodynamic characteristics in it, especially the analysis of patient-specific cases.

Here, we used a system of computational hemodynamics that we validated previously to obtain information on blood flow after the Norwood procedure [4, 5] to analyze the hemodynamic characteristics of a patient-specific mB-T shunt. The local pressure, blood flow distribution and wall shear stress (WSS) were evaluated to predict clinical outcomes. The aim of the study is not only to disclose the hemodynamic characteristics at the area of the mB-T shunt, but also to confirm the applicability of our computational system in the hemodynamic analysis of the mB-T shunt for future surgical optimizations and establishment of quantitative standards to evaluate the outcomes of the mB-T shunt.

II. Materials and Methods

A. Generation of Geometric Models

Clinical studies were done with informed consent of the parents of a 12-month-old child that underwent the mB-T shunt. Protocols were approved by the local institutional review board and regional research ethics committee. To obtain patient-specific computed tomography (CT) images, we used a series of fixed protocols for determining the amount and the density of the imaging contrast agent. A series of continuous 0.625 mm thick CT images of the thorax to

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reconstruct three-dimensional (3D) vascular geometry were recorded by 16-slice multi-detector row enhanced CT scanner (Bright Speed Elite, GE Medical System, General Electric, America). Medical imagining software RealINTAGE® (KGT Co. Tokyo, Japan) compiled and reconstructed the 3D vascular geometry. The accuracy of the reconstructed model had been checked against the geometry with an exacted measurement carried out by the original DICOM CT slice files. A non-shrinking smoothing technique generated a numerical model for CFD simulation [6]. Figure 1 depicts the geometry after surface smoothing. Table 1 lists the dimensions of the arterial geometry and the implanted shunt.

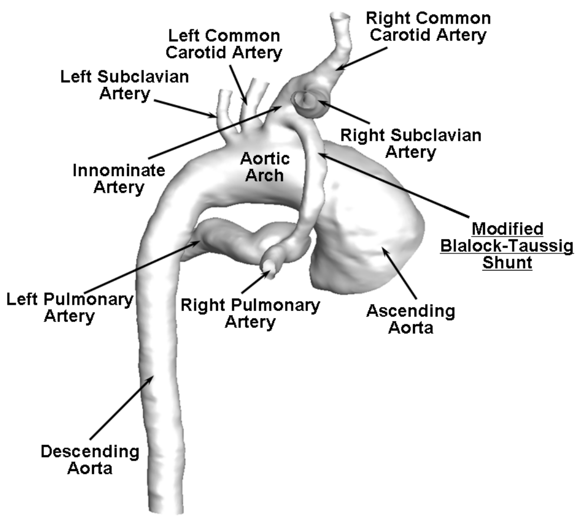

Figure 1 Patient-specific three-dimensional vascular geometry after the modified Blalock-Taussig shunt

Table 1 Dimensions of the arterial geometry and implanted shunt

Artery	Lumen Diameter (mm)
Ascending Aorta	21.87
Right Pulmonary	3.46
Left Pulmonary	6.39
Descending Aorta	6.86
Left Subclavian	2.87
Left Common Carotid	3.33
Right Common Carotid	3.75
Right Subclavian	3.64
B-T shunt	3.98

*Note: All diameters in Table 1 refer to characteristic diameters.

B. Governing Equations

The equations governing blood flow are the 3D incompressible Navier-Stokes equations. The motion of blood flow can be described by the following equations for the conservation of mass and momentum:

Continuity equation

$$\nabla u = 0 \tag{1}$$

Momentum equations

$$\rho \frac{Du}{Dt} = -\nabla p + \mu \nabla^2 u \tag{2}$$

where ρ is the fluid density, u is the velocity vector, p is pressure, μ is dynamic viscosity and t is time. Because arteries are large relative to individual blood cells and shear rates are greater in larger arteries, we assumed blood to be a Newtonian fluid with constant density (ρ = 1060 kg/m^3) and viscosity (μ =4.0×10^{-3} Pa s).

The maximum value of Reynolds Number (Re) was about 4000 at the aortic arch in one cardiac cycle based on our previous studies [5]. An unstable and a strong transition flow occurred from the systole to diastole when the flow decreased the speed to laminar flow in a relatively short period. Therefore, the most widely validated turbulence model; the standard $k-\varepsilon$ model, was employed to solve the complex flows [7]. The $k-\varepsilon$ model is a two-equation eddy viscosity model based on the solution of equations for the kinetic energy of turbulence (Eq. (4)) and the turbulence dissipation rate (Eq. (5)) defined below:

Eddy viscosity

$$\mu_t = \rho C_\mu \frac{k^2}{\varepsilon} \tag{3}$$

Turbulence kinetic energy

$$\rho \frac{\partial k}{\partial t} + \rho u_j \frac{\partial k}{\partial x_j} = \tau_{ij} \frac{\partial u_i}{\partial x_j} - \rho \varepsilon + \frac{\partial}{\partial x_j}\left[\left(\mu + \frac{\mu_t}{\sigma_k}\right)\frac{\partial k}{\partial x_j}\right] \tag{4}$$

Turbulence dissipation rate

$$\rho \frac{\partial \varepsilon}{\partial t} + \rho u_j \frac{\partial \varepsilon}{\partial x_j} = C_{\varepsilon 1} \frac{\varepsilon}{k} \tau_{ij} \frac{\partial u_i}{\partial x_j} - C_{\varepsilon 2} \rho \frac{\varepsilon^2}{k} + \frac{\partial}{\partial x_j}\left[\left(\mu + \frac{\mu_t}{\sigma_\varepsilon}\right)\frac{\partial \varepsilon}{\partial x_j}\right] \tag{5}$$

Closure coefficients

$C_{\varepsilon 1}$=1.44, $C_{\varepsilon 2}$=1.92, C_μ=0.09, σ_k=1.0, σ_ε=1.3

To evaluate the outcomes of the mB-T shunt on the balance of blood flow distribution between sympatric and pulmonary circulations, the flow distribution ratio (FDR) is given by,

$$FDR = \frac{Q_{outlet}}{Q_{inlet}} \times 100\,\% \tag{6}$$

where Q_{outlet} is the flow in the left pulmonary artery (LPA), right pulmonary artery (RPA), descending aorta (DA), left subclavian artery (LSA), right subclavian artery (RSA), left common carotid artery (LCCA), right common carotid artery (RCCA) and Q_{inlet} is the inflow at the ascending aorta (AAo).

Wall shear stress (WSS), thought to be a causal factor for some arterial diseases, is a measure of the interaction between the fluid and its solid boundary. The equation for WSS symbolized by τ_{wall} in a Newtonian fluid is given by:

$$\tau_{wall} = -\mu \left.\frac{\partial u_x}{\partial y}\right|_{y=0} \tag{7}$$

where μ is viscosity, u_x is the velocity of the fluid near the boundary, and y is the height above the boundary. Studies have shown a strong correlation in the magnitude of shear stress, endothelial cell function, and vessel wall remodeling [8]. When WSS is high, the effects of shear stress on the boundary wall must be taken into account. Thus, high WSS is one reason for damage to the endothelial layer of blood vessels. Moreover, low WSS coincided with areas of low flow velocity is considered to promote platelet activation, and it is one of the reasons for causing shunt thrombosis and occlusion [9].

C. *CFD Analysis*

1) Mesh generation

The grid-generation software, ANSYS®-ICEM 13.0, was applied to discretize the computational domain with a combination of tetrahedral grids in the interior and five-layer body-fitted prismatic grids in the near-wall regions to improve the resolution of the relevant scales in fluid motion.

To find the best mesh for CFD analysis, grid-independent verification were performed in our previous study and found that grid numbers of about one million would make the most efficient mesh [5]. In the present study, we used 1,310,008 grids and 396,317 nodes on the patient-specific 3D vascular geometry.

2) Boundary conditions

Due to the pulmonary atresia before performing the mB-T shunt, only the left ventricle provided blood flow to the body through the AAo. The implanted shunt distributed the blood flow from the systemic circulation to pulmonary arteries after the procedure.

To fully develop the flow boundary layer, we extended the inlet domain upstream to twenty times the size of the AAo. The pulsatile mass flow, measured by echocardiography in real-time with an electrocardiogram recorder, was applied as the inlet boundary condition at the extended inlet. Figure 2 shows mass flow in the AAo in one cardiac cycle.

At the outlet, we extended vessel diameter sixty times in a normal direction to allow sufficient recovery of blood pressure in each branch. We assumed there was a zero pressure gradient at the outlets. The backward pressure wave proposed in our previous study [5], which was considered the pressure wave reflection from peripheral vessels at diastolic phase, was used to model cardiovascular flow.

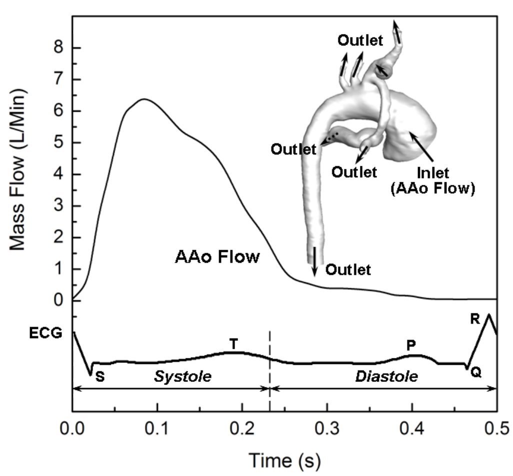

Figure 2 Mass flow in the ascending aorta

3) Calculation

The finite volume solver package ANSYS®-FLUENT 13.0 was applied to solve the physiologic flow in the patient-specific model. For this simulation, we assumed vascular wall consist of rigid surfaces. The semi-implicit (SIMPLE) method was chosen to solve the discretized 3D incompressible N-S equations. We performed a steady calculation at the systolic peak as the initialized state for the pulsatile flow simulation. Details of our method for calculating pulsatile flow were reported previously [4,5]. All the calculations were performed on a computer workstation with a 64 bit Windows 7 operating system. The workstation was equipped with double CPUs: Intel (R) Xeon X5690 3.46 GHz processors with 24.0 GB RAM memory.

III. Results

Figure 3 panels a-c, depict contour plots of the total pressure, WSS, and streamline distribution at the systolic peak. Figure 3a shows high pressure drop in the mB-T shunt, approximately 60 mmHg. Relatively low pressure was occurred at the connection between the shunt and the pulmonary artery. Figure 3b displays relatively high values of WSS in the shunt and the orifice of the mB-T shunt. Figure 3c shows streamlines at the peak of systole. There were high magnitudes of velocity in the mB-T shunt. Blood flow accelerated and became more complex and unstable when it passed through the shunt.

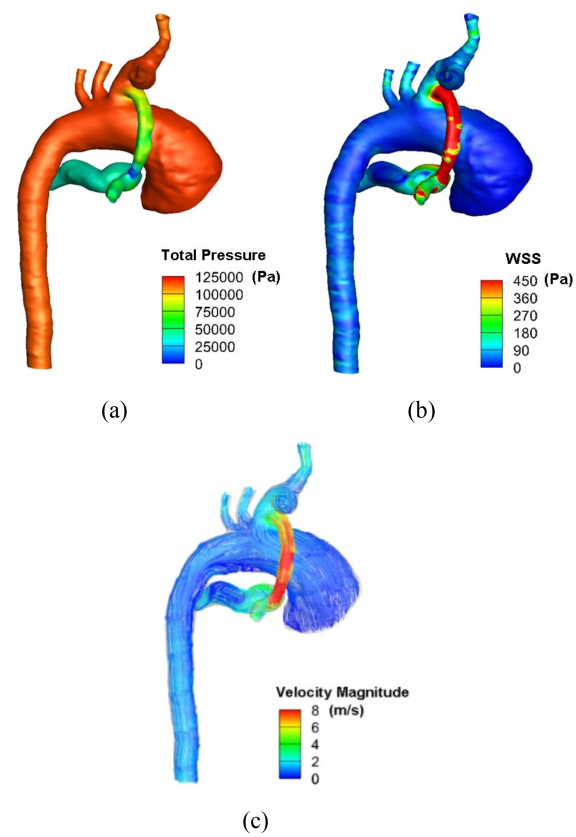

Figure 3 Hemodynamic parameters at the peak of systole

(a) Total Pressure; (b) WSS; (c) Velocity Magnitude

The calculation also revealed the average proportion of total blood flow distributed into each branch vessel during a single cardiac cycle. Seven main branch vessels distributed the flow from the AAo. Table 2 lists the results computed for the average distribution of blood in one cardiac cycle. They were within a physiologically realizable range.

Table 2 Blood distribution in one cardiac cycle

Proportion of Blood Flow (%)						
Vessel	B-T Shunt	DA	LSA	RSA	LCCA	RCCA
Ration	42	45	2.5	4.2	3	3.3

*Note: DA: descending aorta; LSA: left subclavian artery; RSA: right subclavian artery; LCCA: left common carotid artery; RCCS: right common carotid artery.

IV. Discussion

The mB-T shunt should not supply excessive pulmonary blood flow that might result in a low cardiac output and therefore increased the risk of cardiac arrest and death as well as elevated pulmonary vascular resistance [9]. Studies suggest that maximal oxygen delivery occurs at the value of the Q_p/Q_s (where Q_p is the flow distributes to pulmonary arteries through the shunt, and Q_S is the flow distributes to systemic circulation) between 0.5 and 1 [10]. In the present study, the average distribution of the AAo flow through the mB-T shunt is about 42% during one cardiac cycle. The value of Q_p/Q_s is 0.724, which may achieve most appropriate distribution of the blood flow on the viewpoint of oxygen delivery.

Post-operative thrombosis and occlusion of the mB-T conduit can be sudden and fatal especially in the peri-operative period [11]. Low-speed flow area was found near the anastomotic site of the mB-T conduit to the pulmonary artery. Stagnant area in the flow domain and large WSS may both stimulate the function of the platelet which in turn results in the coagulation of the blood and formation of the thrombus. The occlusion of the conduit will lead to severe low blood oxygen saturation during the peri-operative period and often need emergency reoperation to replace a new conduit.

There are most likely two limitations of the present study which should be considered. Firstly, this study used the rigid vessel model in the calculation. Therefore, the interaction of pulsatile flow and vessel compliance were not taken into account. According to our previous studies based on the analysis system, there a little delay of the calculated pressure wave. The maximum differences between the calculated pressure and clinical data measured by a catheter with a pressure sensor were all less than 10%. Secondly, the present study was a single patient-specific research. More cases should be investigated in the following studies.

V. Conclusion

Here we used the system of computational hemodynamics that we validated previously to obtain information on blood flow after the Norwood procedure [4, 5] to analyze the hemodynamic characteristics of the mB-T shunt for the evaluation of patient-specific outcomes. The calculated results were within a physiologically realizable range. There were relatively lager pressure drop in the mB-T conduit. High WSS was generated when the flow distributed to the pulmonary arteries from the conduit. The results suggest pressure and WSS is the key parameters for evaluating the flow in the mB-T shunt. They also suggest the low-speed flow areas and stagnant areas, and flow impingement areas created in the mB-T conduit have major effects on the outcomes of this procedure. The next challenge will be to adopt this approach to optimize the implantation of the mB-T conduit.

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