Evaluation of a novel Y-shaped extracardiac Fontan baffle using computational fluid dynamics

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Objectives: The objective of this work is to evaluate the hemodynamic performance of a new Y-graft modification of the extracardiac conduit Fontan operation. The performance of the Y-graft design is compared to two designs used in current practice: a t-junction connection of the venae cavae and an offset between the inferior and superior venae cavae.

Methods: The proposed design replaces the current tube grafts used to connect the inferior vena cava to the pulmonary arteries with a Y-shaped graft. Y-graft hemodynamics were evaluated at rest and during exercise with a patient-specific model from magnetic resonance imaging data together with computational fluid dynamics. Four clinically motivated performance measures were examined: Fontan pressures, energy efficiency, inferior vena cava flow distribution, and wall shear stress. Two variants of the Y-graft were evaluated: an ‘‘off-the-shelf’’ graft with 9-mm branches and an ‘‘area-preserving’’ graft with 12-mm branches.

Results: Energy efficiency of the 12-mm Y-graft was higher than all other models at rest and during exercise, and the reduction in efficiency from rest to exercise was improved by 38%. Both Y-graft designs reduced superior vena cava pressures during exercise by as much as 5 mm Hg. The Y-graft more equally distributed the inferior vena cava flow to both lungs, whereas the offset design skewed 70% of the flow to the left lung. The 12-mm graft resulted in slightly larger regions of low wall shear stress than other models; however, minimum shear stress values were similar.

Conclusions: The area-preserving 12-mm Y-graft is a promising modification of the Fontan procedure that should be clinically evaluated. Further work is needed to correlate our performance metrics with clinical outcomes, including exercise intolerance, incidence of protein-losing enteropathy, and thrombus formation.

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Supplemental material is available online.

Computational modeling, clinical observation, and previous experimental work suggest the geometry of the total cavopulmonary connection (TCPC; Fontan) plays a key role in energy losses,[1-8] that is, efficiency, and Fontan outcomes. Previous simulation-derived alternatives led to the adoption of an offset of the inferior anastomosis relative to the superior anastomosis and reduced energy losses when compared with the traditional t-shaped junction.[3,9,10] Recent work has also examined the effects of exercise on energy loss in multiple patient-specific models.[11,12]

Recently, and concurrently with our work, Soerensen and associates[13] proposed a similar design called the ‘‘OptiFlo’’ in which the native inferior (IVC) and superior venae cavae (SVC) are bifurcated before the pulmonary artery anastomosis. Their study demonstrated reduced power loss at rest and simulated exercise with simplified geometric models and steady inflow conditions. The two studies share the concept of eliminating flow competition by bifurcating the graft used in the extracardiac Fontan procedure. However, our study addresses some of the computational limitations of this previous study through the use of more detailed geometric models, pulsatile inflow conditions, and sophisticated outflow boundary conditions. In addition, our proposed Y-graft design offers several important technical advantages: it can be easily modified for an individual patient, it can be custom manufactured of synthetic material using realistic sizes, and it allows the procedure to be performed without cardiopulmonary bypass.

We[11] have previously reported the development of an increasingly accurate Fontan modeling system, producing realistic pressure and flow data when compared with that measured by catheterization and echocardiography. This was accomplished by incorporating increased anatomic accuracy, specifically a greater level of pulmonary branching,

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Abbreviations and Acronyms

IVC	= inferior vena cava
LPA	= left pulmonary artery
MRI	= magnetic resonance imaging
PC	= phase-contrast
RPA	= right pulmonary artery
SVC	= superior vena cava
TCPC	= total cavopulmonary connection
WSS	= wall shear stress

and the effects of respiration and cardiac contraction. The simulation framework used in this work produces results including energy efficiency, absolute pressures, pressure gradients, wall shear stress (WSS), and flow distribution. Several important features of this software have been customized for cardiovascular applications and are not available in standard commercial codes. These include resistance, impedance and lumped model boundary conditions to accurately model physiologic pressure levels,[14,15] Lagrangian particle-tracking methods, wall deformability (not used in this work),[16] and adaptive meshing capabilities.[17]

Results from these patient-specific simulations suggest that despite seemingly excellent technical and hemodynamic results with current clinical markers and diagnostic modalities, marked differences in individual Fontan efficiencies exist. Additionally, during simulated exercise, efficiencies drop substantially, and differences in performance between geometries become more pronounced. These previous results demonstrated that studies focusing solely on resting conditions (and reporting resting efficiencies of 90% based on the ratio of energy outflow to inflow) were not comprehensive representations of the ''physiologic realities'' of the Fontan circulation and provided a strong impetus for increased study and design of more efficient Fontan geometries.

In this study, we describe and evaluate a new modification of the extracardiac Fontan procedure using computer-aided design and computational fluid dynamics. This modification incorporates a Y-shaped graft to replace the cylindrical polytetrafluoroethylene tube grafts (Gore-Tex; W. L. Gore & Associates, Inc, Flagstaff, Ariz) currently used to connect the IVC to the pulmonary arteries. We demonstrate that the Y-graft results in higher energy efficiency and more equal distribution of IVC flow to both left and right lungs when compared with current operative techniques.

METHODS

Designing new surgical techniques, that is, virtual surgery, requires image data acquisition, construction of patient-specific models, fluid mechanics simulations, and post-processing of results.[18] In this work, we compare the performance of the proposed design to current TCPC designs using several clinically motivated metrics of performance obtained from post-processing our simulation results. The methods for each of these steps are outlined below.

Image/Anatomy Acquisition

Magnetic resonance angiography was performed with a 1.5-T magnetic resonance imaging (MRI) scanner (Signa TwinSpeed; General Electrics, Milwaukee, Wis). During intravenous administration of a gadolinium-based contrast agent, images were acquired using a half-Fourier, 3-dimensional fast gradient–recalled echo sequence with breath-holding. The spatial resolution was approximately 0.7 mm × 1.2 mm × 2.0 mm. Flow waveforms in the venae cavae were recorded with a 2-dimensional phase-contrast (PC) MRI method. The imaging plane was placed perpendicular to the dominant flow with velocity encoding parallel to the flow. The slice thickness was 10 mm and the in-plane resolution was about 1.0 mm × 1.7 mm. The encoding velocity was 120 cm/s. Velocity data were acquired over several cycles of free breathing with the use of cardiac gating and respiratory compensation.

Model Construction

Once the image data have been acquired, there are four steps necessary to construct geometric models from image data volumes (Figure 1)[19] (1) Centerline paths are created in the vessels of interest, (2) segmentations of the vessel lumen are created perpendicular to the centerlines using a 2-dimensional level set method, (3) the 2-dimensional segmentations are lofted together, creating a solid model of the desired vasculature, and (4) the solid model is discretized into an unstructured tetrahedron mesh for use in the finite element flow solver.

For this study, one patient-specific anatomy and three geometric variations were constructed. The first model was constructed from patient-specific MRI data from a 4-year-old girl who had a traditional extracardiac Fontan with the IVC and SVC anastomosed to the pulmonary arteries in the classic ''t-junction'' configuration (that is, with no offset, Figure 2). Subsequently, by use of the same custom model construction software, three variations (left pulmonary artery [LPA] offset, small-Y, large-Y, Figure 2) on the initial, patient-specific design were created by changing the geometry of the IVC connection while keeping the SVC and pulmonary geometries identical. In this way, the effect of the IVC connection geometry on hemodynamic performance could be examined independently and compared with the original t-junction design. The pulmonary vasculature was constructed to include all pulmonary branches larger than or equal to the segmental branches, representing the resolution limits of the MRI data for this patient.

In the offset model, the IVC was offset approximately one IVC diameter from the SVC toward the LPA to represent current clinical practice. Two variations of the Y-graft were constructed.

The first (small-Y) was constructed with an 18-mm trunk with two 9-mm branches, modeled after grafts that are approved by the Food and Drug Administration and currently available. The second (large-Y) was designed to approximately preserve the cross-sectional area between the trunk (18 mm) and two branches (each 12 mm). While an exact area preservation would result in 12.7-mm branches, the smaller size was chosen to achieve a better fit to the pulmonary arteries for this patient and because grafts are typically manufactured with standardized integer dimensions. The IVC trunk dimension of 18 mm was chosen in accordance with current practice at our institution. In a recent study of more than 300 patients, 97% received conduits of 18 mm or larger, and 69% received conduits of 20 mm or larger.[9] These results also agree with previous studies determining optimal graft size.[20]

Flow Simulations and Boundary Conditions

To simulate blood flow, we used a custom stabilized finite element solver to solve the time-dependent, 3-dimensional Navier–Stokes equations.[21,22] An anisotropic adaptive meshing scheme[17] was used to ensure mesh convergence using approximately 1.5 million elements for each model. A newtonian approximation for the viscosity was assumed with a value of 0.04 g/(cm s) and the density of blood was 1.06 g/cm^3. A rigid-wall approximation was used. Further details on our Fontan simulation methods and flow solver can be found in the article by Marsden and associates.[11]

Respiratory-varying boundary conditions were imposed at the IVC inlet in all simulations. Pulsatile flow rates were acquired from patient-specific

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