Surgical Optimization of the Glenn and Fontan Procedures using a Y-Shaped Graft Extension for Patients with Hypoplastic Left Heart Syndrome

Introduction

Congenital heart defects (CHD) are serious forms of structural deformities that arise from abnormal formation of the heart and its major blood vessels. According to the American Heart Association, congenital heart defects are the number one cause of death from overall birth defects [1]. One of the most common forms of CHD is known as hypoplastic left heart syndrome (HLHS), where the left side of the heart is critically underdeveloped. Recent surgical advances have allowed for a procedure known as three-stage palliation, where the right, fully-functioning side of the heart is converted into the main systemic blood flow pump to the rest of the body. This three-stage procedure, namely the Norwood, Glenn and Fontan stages, is widely considered a huge medical success, due to low mortality rates following the procedure. However, serious clinical challenges remain despite post-operative survival rates of up to 96% [2], such as diminished exercise capacity, protein losing enteropathy, arteriovenous malfunction, thrombosis, arrhythmias and overall heart failure [3]. Therefore, surgical optimization of this procedure to improve overall patient outcomes is evident.

Reduced exercise performance for post-Fontan patients is well documented, as exercise can only be responded to by increasing heart rate due to limited stroke volume [4]. Sundareswaran and coworkers used a computational model to correlate Fontan graft geometry with cardiac function and suggested that optimizing graft geometry may improve exercise capacity for post-Fontan patients through a reduction in the pressure difference between the vena cava and left atrium [5]. Furthermore, animal models by Guyton et. al [9] showed that reducing direct flow collision and optimizing flow distribution in the area of anastomosis has a profound effect on overall energy efficiency. Therefore, it is hypothesized that minimizing direct flow collision and decreasing flow resistance at the connection area may help lower central lower venous pressure, resulting in an increase in energy efficiency and consequently, increased exercise capacity.

Various optimizations have been proposed to improve flow distribution and venous resistance in current total cardiopulmonary connections (TCPC). Ensley et al. proposed an offset TCPC design as shown in Figure 1, resulting in a 68% reduction in energy loss [10].

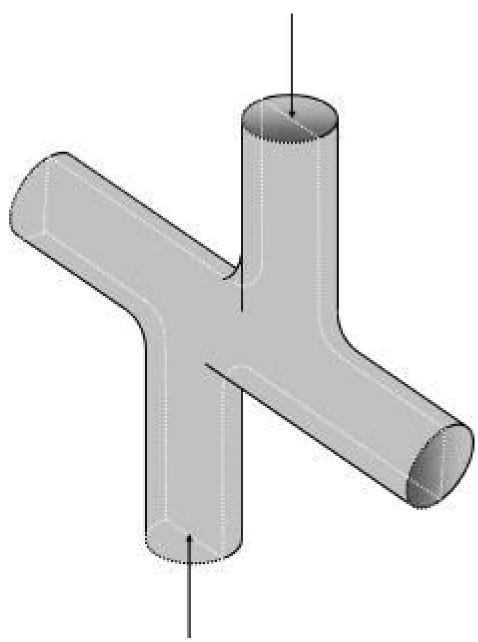

Figure 1: Offset TCPC Design as Proposed by Ensley and co-workers

However, this implementation produced uneven flow distribution due to increased venous resistance in the right pulmonary artery relative to the left pulmonary artery, resulting in the formation of pulmonary arteriovenous malformations. Bifurcation TCPC designs have been proposed by Yang and co-workers, shown in Figure 2, to reduce direct flow collision in these models, resulting in a 45% reduction in energy loss [3]. However, the design not only significantly increased downstream venous resistance, but also increased surgical technical difficulty.

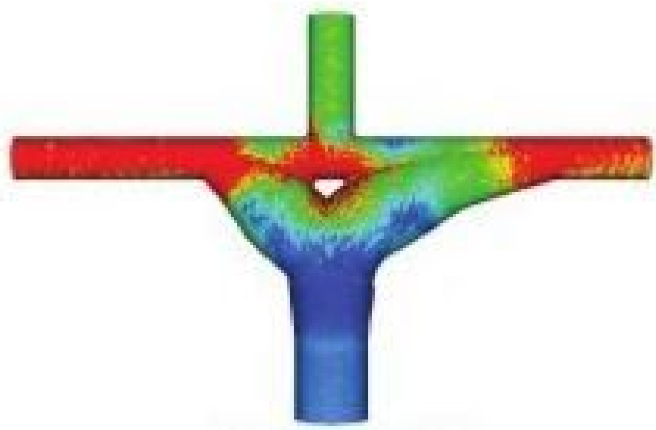

Figure 2: Bifurcation IVC TCPC Design as Proposed by Wang and co-workers

A novel Y-shaped graft has been proposed by Soerensen and co-workers [12], shown in Figure 3, with the effort of reducing direct flow collision in the area of anastomosis by introducing a curvature at the ends of the graft connections, achieved through a Y- shaped geometric variant of the graft attachment forming the TCPC connection.

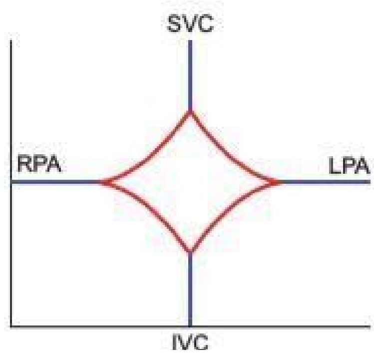

Figure 3: Novel Y-Shaped Graft Design of Superior and Inferior Vena Cavas

Computational analysis on the Y-shaped graft extension for the inferior vena cava has been performed by Santhanakrishnan and co-workers [6], where venous pressure was decreased by 5 - 10 mmHg and overall energy loss was decreased by 20 - 50% from the normal TCPC model. However, no studies have been performed on the Y-shaped design of both the superior and inferior vena cavas independently, nor on the extent of the graft variant that can improve energy efficiency and minimize direct flow collision.

In this study, this optimization is analyzed with respect to the curvature radius for optimal graft design of the anastomosis, and inspected with efficiency and performance metrics, namely pressure and energy loss. It was hypothesized that, due to Poiseuille's equation, venous resistance would decrease with increasing curvature radius and, therefore, pressure values would decrease in the area of anastomosis. Furthermore, due to increased area in the end-graft region, streamlines would converge at a flatter angle, resulting in a reduction in direct flow collision, and consequently, a reduction in energy loss. Optimizing the aforementioned parameters in this manner will decrease the energy required for passive blood flow to the lungs via the pulmonary circulation, decreasing the load on the heart, and increasing exercise capacity in post-Fontan patients.

Methods

SolidWorks was used to create and simulate flow conditions on a series of models of total cavopulmonary connections. Model values were obtained through in-vivo venous measurements obtained by Santhanakrishnan and co-workers [6]. Vessel lengths were specified at 80 mm for the superior vena cava, 140 mm at the inferior vena cava, and each pulmonary artery 80 mm. Internal wall diameter and wall thickness were set at 20mm and 1mm, respectively. A control TCPC model was created with the aforementioned values composed of hard edges at the connection area. Y-shaped graft models were composed of curvature variations at the SVC and IVC, evaluated independently, with measurements of 10 mm, 20 mm and 30 mm.

In the case of wall boundary assumptions, Bazilevs et al. applied the state of art fluid structure interaction to a patient specific Fontan model and showed that pressure was overestimated in the simulation of rigid walls [7]. However, recently, Long et al performed Fontan FSI simulations with variable wall properties and showed that the difference in energy loss and flow distribution were small [8]. Therefore, a rigid wall assumption is still adequate to evaluate Fontan patients' hemodynamic performance and was used in the simulation.

Flow pulsatility in Fontan patients is reduced as the inferior and superior vena cavas are disconnected from the right atrium. Due to the lack of pulsatile flow, steady flow was considered in the simulation. Flow rates were comparable to the superior vena cava average flow rate, 28.5 cm/s, and inferior vena cava average flow rate, 20.3 cm/s, over one cardiac cycle, as determined by Foust through in-vivo measurements [11]. Blood was assumed incompressible and non-Newtonian, and a no-slip boundary condition was imposed on the walls. Ambient air pressure of $1x10^6$ dynes/cm^2 is applied at the outlet due to assumed negligible absolute pressure values in the venous system, and effect of gravity was neglected.

Geometric variants were analyzed with respect to flow distribution, pressure and energy loss in the area of anastomosis and along the curvature radii. Energy loss was computed through the head loss calculation, shown in Eq 1, which computes the head loss of a general pipe due to bend. Values between models for surface roughness, friction factor and viscosity were held constant, while maximum velocity and the K factor were variable. Average velocity measurements were taken at the curvature radius of each model. The value of K can be calculated from the geometry of the end-graft based on empirical results obtained from a variety of test data. Correspondingly, the K values for the control TCPC model and both SVC and IVC variations of 10mm, 20mm and 30mm were 1.0, 0.91, 0.83, 0.77, respectively [14].

$$hL= Kv^2/2g,$$

Results

Models were computed using SolidWorks Flow Analysis Toolbox and analyzed using velocity flow trajectories and pressure surface plots, as shown in Figure 4 and 5. Figure 4 shows an overall decrease in pressure in both SVC and IVC Y-graft optimizations as opposed to the control in the area of anastomosis. This pressure decrease was more pronounced in SVC graft variations than IVC graft variations. No change in pressure was observed in the area of anastomosis in geometric variants of IVC and SVC models. However, upstream pressure in the SVC and IVC was significantly decreased with increasing curvature radii.

Figure 5 shows a reduction in direct flow collision with increasing curvature radii, especially through inspection of streamlines guided by each curvature. Furthermore, streamline convergence was enhanced with increasing curvature radius, improving overall flow distribution. Absolute velocity measurements taken at each curvature in SVC variations (A, B, C) yielded 11.25%, 22.72% and 30.38% decreases, respectively, compared to the control. IVC model variations (models D, E, F) yielded 15.17%, 31.81% and 69.62% decreases in energy loss, respectively.

Discussion

The overall decrease in pressure of both the SVC and IVC models in the area of anastomosis relative to the control model observed in the Y-shaped optimization is corroborated by Santhanakrishnan and co-workers [6], who observed a 5 -10 mmHg difference in connection area pressure following placement of the unidirectional Y-shaped valve. Furthermore, a decrease in pressure along the walls of each curvature was observed with increasing curvature radii.These decreases in pressure are hypothesized to be due the overall increase in cross-sectional area with the inclusion of the graft optimization. However, differences in maximum pressure in the area of anastomosis with increasing curvature radii were not observed. It is hypothesized this may be due to direct flow collision within the middle of the connection area. Therefore, optimization of the graft connection will result in static maximum pressure values from model to model due to collision of the streamlines in the middle of the connection area that are not fully optimized, despite variations in end-graft area. Furthermore, this pressure decrease, relative to the control TCPC model, is more pronounced in the SVC than the IVC models. It is hypothesized this may be due to the inherent flow characteristics of the SVC and the IVC, specifically where the SVC is flowing directly downward at a relatively greater velocity than the IVC, while blood from the IVC is flowing upward. Therefore, optimization of the SVC yields slightly lower pressure values in the connection area relative to the IVC optimizations.

While maximum pressures in the connection area between model variations were relatively constant, upstream pressures of the optimized vein were significantly reduced with increasing curvature radii. Again, this may be due to increased graft area with each optimization, increasing the pressure gradient between the SVC or IVC and the connection area, enhancing overall flow rate characteristics.

Flow distribution was significantly enhanced with increasing curvature radii, as streamlines were guided by the curvature radius, reducing direct flow collision. Again, this is corroborated by Santhanakrishnan and co-workers [6], who found a 20-50% decrease in energy loss relative to the control TCPC model mainly due to enhanced flow distribution. Furthermore, Zhang and co-workers examined examined local hemodynamics for patients with pulmonary arterial hypertension, and found the radius of curvature had a profound effect on flow stability and energy efficiency [13].As such, an overall decrease in energy loss was observed with increasing curvature radii of each model. A notable profound decrease in energy loss was observed in the IVC 30mm variation as opposed to the SVC 30mm variation. Again, this could be due to the inherent flow characteristics of the vessel, and may suggest a focus on IVC optimizations for future work. Overall, while SVC optimizations yield lower pressure values and similar overall energy losses at the onset, IVC optimizations may have a greater area for potential.

Flow analysis in this study was severely limited to the application of one boundary condition per inlet/outlet, as well as the steady-state boundary conditions applied to the outlets. Yang et. al suggest that applying constant pressure at the outlets may alter flow field and pressure distribution. Therefore, time-dependent pressure and flow rates corresponding to physiological boundary conditions are typically preferred [6]. Furthermore, ambient air pressure at the outlets was required in SolidWorks due to the simulation solver treating the vessel as a mechanical part. Therefore, when pressure was reduced to zero, absolute pressure within the vessel was negative, suggesting a calculation of the weight of the fluid. Furthermore, Hsia and co-workers found that the inclusion of gravity had a profound effect on infra-diaphragmatic venous flow. Simulated models in this study are still valid strictly for comparative purposes between SVC and IVC variations relative to the control model. However, further testing is needed to determine absolute values with the inclusion of time-dependent boundary conditions as well as gravity. Furthermore, increased resolution may be required to definitively inspect streamlines and analyze the effect of curvature radius on the flow distribution.

Further research is needed to determine the best TCPC optimized implementation for post-Fontan patients. While the 30mm curvature radius IVC Y-shaped model did yield the greatest reduction in energy loss compared to the bifurcated and offset models, future work is needed to include time-dependent inlet and outlet conditions to determine if these results are significant. Furthermore, the Y-shaped optimization of both the IVC and SVC at the same time should be inspected to determine the greatest optimization for this procedure. A combination of the aforementioned models can also be employed to analyze the most efficient model for implantation in Fontan patients with the effort of increasing exercise capacity.

Similarly, time-dependent boundary conditions should be applied to simulate physiological conditions with the greatest accuracy. The inclusion of gravity should be considered, as well as the effect of respiration; Qian et. performed energy loss analysis on patient-specific TCPC models using MRIs and found that respiration had a profound effect on overall energy loss, with as much as a 1.5x difference between simulations that included respiration and simulations that did not [16]. The Y-shaped optimization should be inspected in greater detail, specifically the extent to which the curvature radius can be increased before insufficient pressure for optimal flow into the pulmonary artery occurs.

Conclusion

The effect of variations of the Y-shaped graft implementation on the SVC and IVC for post-Fontan patients was investigated and compared to the normal TCPC surgical implementation. Both models yielded a relative decrease in connection area pressure, with variations in curvature radius yielding decreased upstream venous pressure and decreased pressure at the curvature radius. Furthermore, direct flow collision and energy loss were significantly reduced with increasing curvature radii. Optimization of these parameters, specifically venous resistance and direct flow collision, has been shown to increase overall energy efficiency, which may lead to decreased load on the heart, increased peak oxygen metabolic intake and enhanced overall flow characteristics, which may lead to increased exercise capacity. By exercising exercise capacity, post-surgical outcomes can be improved for Fontan patients, leading to an increase in the quality of life.

Future work should analyze the Y-shaped graft implementation on both the IVC and the SVC at the same time, as well as a combination of the offset, Y-shaped and bifurcated models. Time-dependent inlet and outlet conditions should be employed, gravity should be included and resolution should be increased to determine the overall effect of curvature radius on flow distribution and the minimization of direct flow collision.

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Appendix

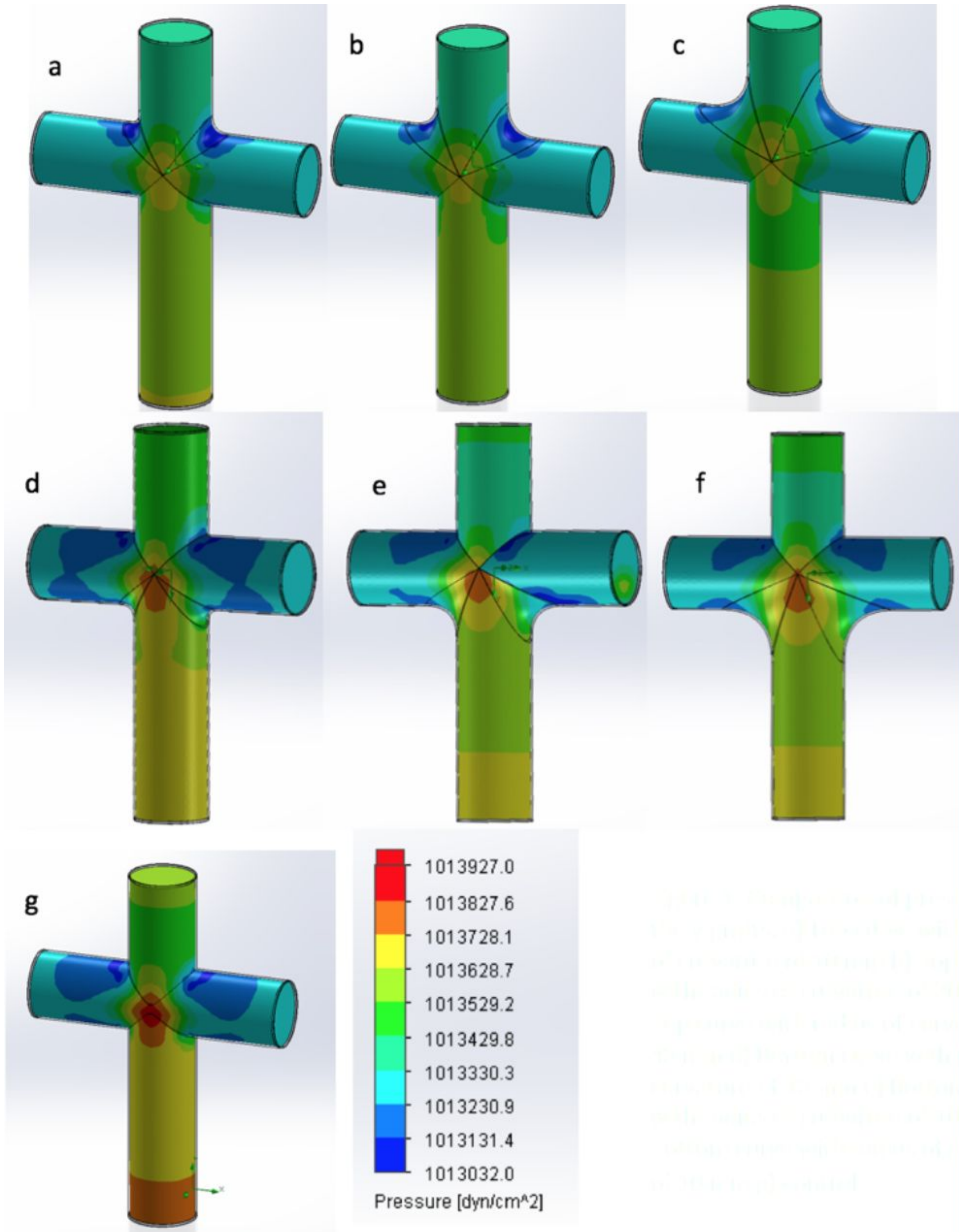

Figure 4: Pressure contour maps in superior vena cava (SVC) and inferior vena cava radius of curvature variations. a. SVC 10 mm radius of curvature b. SVC 20 mm radius of curvature c. SVC 30 mm radius of curvature d. IVC 10 mm radius of curvature e. IVC 20 mm radius of curvature f. IVC 30 mm radius of curvature g. Control

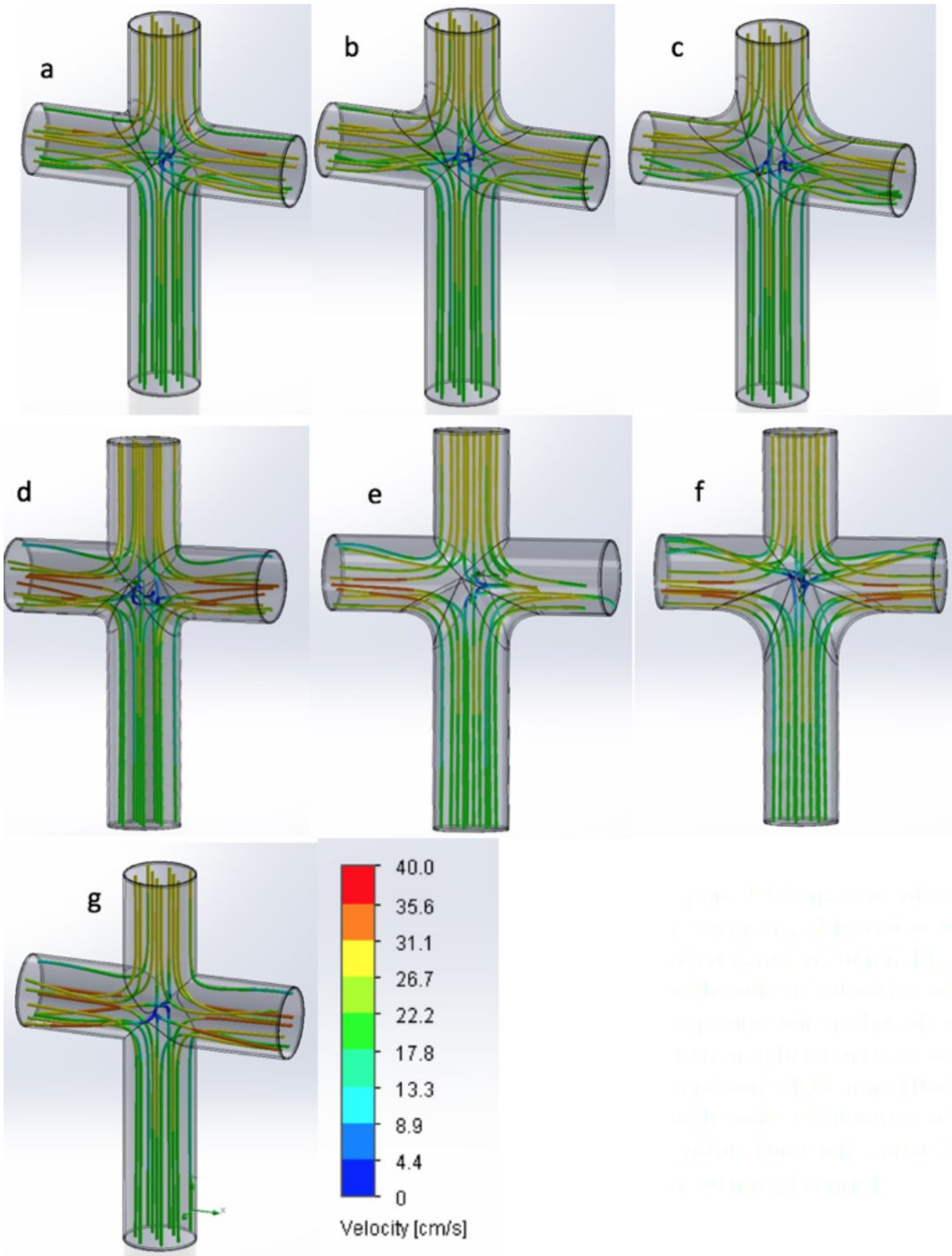

Figure 5:Velocity flow trajectory profiles in superior vena cava (SVC) and inferior vena cava radius of curvature variations. a. SVC 10 mm radius of curvature b. SVC 20 mm radius of curvature c. SVC 30 mm radius of curvature d. IVC 10 mm radius of curvature e. IVC 20 mm radius of curvature f. IVC 30 mm radius of curvature g. Control