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Experimental Investigation of the Effect of Non-Newtonian Behavior of Blood Flow in the Fontan Circulation

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Abstract

The Fontan procedure for univentricular heart defects creates a unique circulation where all pulmonary blood flow is passively supplied directly from systemic veins. Computational simulations, aimed at optimizing the surgery, have assumed blood to be a Newtonian fluid without evaluating the potential error introduced by this assumption. We compared flow behavior between a non-Newtonian blood analog (0.04% xanthan gum) and a control Newtonian fluid (45% glycerol) in a simplified model of the Fontan circulation. Particle image velocimetry was used to examine flow behavior at two different cardiac outputs and two caval blood flow distributions. Pressure and flow rates were measured at each inlet and outlet. Velocity, shear strain, and shear stress maps were derived from velocity data. Power loss was calculated from pressure, flow, and velocity data. Power loss was increased in all test conditions with xanthan gum vs. glycerol (mean $10 \pm 2.9\%$ vs. $5.6 \pm 1.3\%$, $p=0.032$). Pulmonary blood flow distribution differed in all conditions, more so at low cardiac output. Caval blood flow mixing patterns and shear stress were also qualitatively different between the solutions in all conditions. We conclude that assuming blood to be a Newtonian fluid introduces considerable error into simulations of the Fontan circulation, where low-shear flow predominates.

Keywords

congenital heart disease; Fontan; biofluid dynamics; non-Newtonian fluid; wall shear stress

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1. Introduction

Blood is a complex two-phase fluid of formed elements (such as red blood cells and platelets) suspended in plasma, an aqueous solution of proteins, salts, and organic molecules [1]. While blood is a non-Newtonian fluid with shear-thinning behavior (i.e. its apparent viscosity decreases as shear rate is increased) [1], its flow characteristics have been observed to approximate Newtonian fluid behavior in large arteries [2]. As such, the assumption of blood as a Newtonian fluid is common in computational and in vitro experimental studies of cardiovascular biofluid mechanics. In general, the non-Newtonian properties of blood are only apparent at shear rates less than 100 s^{-1} [3]. However, non-Newtonian behavior has been shown to have a significant influence in the microvasculature [4] and in a variety of settings where low-shear flow is more prevalent [5-7]. In particular, non-Newtonian behavior can affect velocity profile blunting, flow separation, energy dissipation, and magnitude and distribution of wall shear stress [4,8,9]. Multiple blood rheology models have been utilized in numerical and computational simulations including Casson, Carreau-Yasuda, K-L, power law, and Lattice Boltzmann models to better incorporate these non-Newtonian flow characteristics [10].

In the cardiovascular system, wall shear stress plays an important role in maintaining normal blood vessel function and growth [2]. Normal levels of arterial wall shear stress ($10\text{-}70 \text{ dyne/cm}^2$) promote endothelial quiescence and expression of atheroprotective mediators, while abnormally low wall shear stress ($<4 \text{ dyne/cm}^2$) leads to endothelial proliferation and production of atherogenic substances [11]. Generally, the normal wall shear stress range in arteries is an order of magnitude larger than that in veins [11]. However, in patients who have undergone the Fontan procedure, the final stage of surgical palliation for univentricular congenital heart defects, blood from the central systemic veins (superior vena cava [SVC] and inferior vena cava [IVC]) bypasses the heart and directly flows into the pulmonary arteries [12]. Therefore, by its very design the Fontan procedure exposes the pulmonary arterial bed to abnormally low wall shear stress. This chronic non-pulsatile pulmonary blood flow has been shown to lead to endothelial dysfunction in these patients by altering the responsiveness of endothelial nitric oxide synthase [13,14].

Although the Fontan procedure has significantly improved early survival, these patients are still at risk for early mortality as well as severe long-term dysfunction of the heart, lungs, liver, and intestines [15,16]. Therefore, significant research efforts have been made to try to optimize the surgery in order to minimize these complications. In large part due to experimental models demonstrating the energetic efficiency of various Fontan configurations, the Fontan procedure has evolved over the last three decades to the current extracardiac conduit and lateral tunnel techniques [17]. While technological advances have allowed for the development of more sophisticated models over time, for simplification computational models have universally assumed blood to be a Newtonian fluid [18–21] and experimental models have used Newtonian fluids [22–24] without any evaluation of the potential error introduced by this assumption. Extrapolating from prior investigations of non-Newtonian behavior in blood vessels [4,5,7–9], treating blood as a Newtonian fluid could result in incorrect conclusions about power loss, pulmonary blood flow distribution,

and distribution of wall shear stress. Despite this, a few centers have started to use computational fluid dynamics with a Newtonian assumption to design an optimal Fontan conduit for clinical use [19,25]. Although the performance metrics predicted by computational models were promising, the clinical success of these Fontan conduits unfortunately has been limited so far [19,26].

The goal of this investigation was to evaluate how a Newtonian fluid assumption affects flow in an experimental model of the Fontan circulation. We hypothesized that in comparison to a Newtonian fluid, a non-Newtonian fluid would demonstrate different flow profiles, decreased hydraulic efficiency, decreased mixing of inlet flow streams, and larger areas of low wall shear stress.

2. Methods

2.1 Fluids

A solution of xanthan gum in water (XG) was used as the experimental non-Newtonian fluid. Dynamic viscosity was measured using a DV2T cone/plate viscometer (Brookfield AMETEK Inc., Middleboro, MA) at 37°C at 12 shear rates in the range of 1-1000 s⁻¹. A concentration of 0.04% by weight was determined to have similar viscosity to a typical Fontan patient [27]. For interpolation of viscosity between data points for shear rates of 0-30 s⁻¹, viscosity was determined by a Herschel-Bulkley fluid model [28]:

$$\mu = k\dot{\gamma}^{n-1} + \frac{\tau_0}{\dot{\gamma}} \quad (1)$$

where τ_0 is the yield stress, γ (x00307) is the shear strain, and k and n are constants with k = 8.9721 cP·s⁻¹, n = 0.8601, and τ_0 = 20 mPa. For shear rates of 30-300 s⁻¹, viscosity was determined by a Casson fluid model [28]:

$$\mu = \left[\sqrt{\tau_0 \left(\frac{1 - e^{-m\dot{\gamma}}}{\dot{\gamma}} \right)} + \sqrt{\mu_\infty} \right]^2 \quad (2)$$

where μ_∞ is the Newtonian viscosity, m is a constant, and other variables are defined as above with τ_0 = 0.06 dyne/cm², μ_∞ = 0.035 dyne/cm, and m = 100 s. For shear rates greater than 300 s⁻¹, viscosity was modeled as Newtonian at 3.5 cP [28]. The viscosity curve of the XG solution is illustrated in Figure 1. The density of the XG solution was 913.3 kg/m³.

A solution of 45% glycerol in water by volume was used as a control Newtonian fluid with viscosity of 3.5 cP. The density of the glycerol solution was 1099.3 kg/m³. For the PIV measurements, both fluids were seeded with 10-45 μm fluorescent polyethylene microspheres with 532 nm excitation and 605 nm emission peaks (Cospheric LLC, Santa Barbara, CA).

2.2 Fontan Circuit Model

A model of the vena cavae and pulmonary arteries was custom-made from glass as illustrated in Figure 2. Model dimensions were based on prior similar in vitro studies [22,29]. Figure 3 shows a schematic of the mock Fontan circuit. The glass model was submerged in either XG or glycerol, depending on the experimental condition, and connected to the rest of the circuit with rigid tubing. Resistance in each limb was controlled with adjustable external clamps. Steady flow was supplied from the reservoir into the circuit using a S-M0T100 centrifugal pump (Orqis Medical, Lake Forest, CA) controlled with a Levitronix LUI flow controller (Levitronix GmbH, Zurich, Switzerland).

The glass model was illuminated using a Gemini PIV 200 Nd:Yag laser (New Wave Research, Fremont, CA) pulsed at 14 Hz. Time intervals for the particle exposures ranged from 450 ps and 5 ms. Images were recorded using a 2 megapixel, monochrome camera (IGV-B1920M-KC000, Imperx, Boca Raton, FL). Pressure was measured at each inlet and outlet, with luer-connected sensors (26PCBFA6D, Honeywell International, Morristown, NJ).

Flow rates were measured with Transonic inline flow meters (Transonic Systems Inc., Ithaca, NY). Each flow sensor was externally calibrated to each test fluid prior to the experiment using timed fluid collection. Fluid was pumped into a collection vessel using a constant flow pump at rates ranging from 0.1 to 3.8 L/min. A calibration slope was then calculated by comparing flow rates measured by the sensor with calculated flow rates. Total flow into the circuit was set at either 1.5 L/min (low cardiac output) or 2.5 L/min (high cardiac output), corresponding to indexed flows of 2.1 L/min/m² and 3.6 L/min/m² respectively based on a patient body surface area of 0.7 m² estimated from the model dimensions. Flow ratio between the inlet limbs was adjusted by changing resistance to either an SVC/IVC split of 60/40, to simulate higher SVC flow in infants and young children, or 50/50 to simulate older children and adults. This yielded a total of 4 experimental conditions.

2.3 Data Postprocessing

Raw images were processed using PIVview (PIVTec GmbH, Gottingen, Germany). A min/max filter was used to remove outlying data points. Multi-grid interrogation was also utilized. Data from 50-55 image pairs was averaged for each experimental condition. Spatial resolution was 0.8 × 0.8 mm. Velocity and shear strain were calculated using PIVview. Shear strain ($\gamma(x00307)$) was defined as:

$$\dot{\gamma} = \frac{\delta u}{\delta y} + \frac{\delta v}{\delta x} \quad (3)$$

where u and v are the velocity components in the x-axis and y-axis respectively.

In-house code written in MATLAB (MathWorks Inc., Natick, MA) was used to plot streamlines and to determine shear stress, power loss, and Reynolds number. Shear stress (τ) was calculated as:

$$\tau = \mu \dot{\gamma} \quad (4)$$

where μ is the dynamic viscosity as discussed above. Power loss was calculated as:

$$\text{Power Loss} = \sum_{\text{Inlets}} Q(P + \frac{1}{2}\rho v^2) - \sum_{\text{Outlets}} Q(P + \frac{1}{2}\rho v^2) \quad (5)$$

where Q is the volumetric flow, P is the pressure, ρ is the fluid density, and v is the cross-section averaged velocity [22]. Reynolds number (Re) was calculated as:

$$Re = \frac{\rho Q D}{\mu A} \quad (6)$$

where D is the vessel diameter, A is the vessel cross-sectional area, and the other parameters are as above. Q/A was used rather than velocity to minimize any error introduced from velocity calculations. For XG, viscosity was estimated using the average shear rate in each vessel. Re was less than 1000 in all regions of the circuit under all experimental conditions, suggesting laminar flow. As shown in Table 1, however, Re was consistently higher on average by a factor of two with glycerol compared to XG. This was primarily due to the viscosity of XG being on average about two times higher than glycerol.

2.4 Statistics

All variables are reported as mean $\pm$ SD. Mean values of all variables were compared between XG and glycerol using two-tailed Student t-test with $\alpha < 0.05$ set as the threshold for significance.

3. Results

3.1 Streamlines and Velocity Distribution

Figure 4 compares streamlines between glycerol and XG under the four test conditions. Regardless of solution, the majority of flow from the SVC was directed to the RPA, while the majority of flow from the IVC was directed to the LPA. Significant differences were seen in the central area of stagnation. Under the SVC/IVC 50/50 high cardiac output condition, there was a recirculation adjacent to the main SVC flow stream seen only with XG. Conversely, under both of the SVC 60/40 conditions, there was a large recirculation adjacent to the main IVC flow stream seen only with glycerol.

Velocity distributions are illustrated in Figure 5. The overall velocity profiles observed in the outlet limbs were similar between the two solutions, although more blunted with XG. In all conditions there was a large area of stagnation (velocity < 20% of mean) in the center of the circuit between the SVC and IVC.

3.2 Shear Strain

The general shear strain distribution was similar between solutions under all conditions as shown in Figure 6. The average and maximum shear strain magnitude was generally higher with glycerol (mean for all conditions $34.1 \pm 25.3 \text{ s}^{-1}$ for glycerol vs. $25.9 \pm 23 \text{ s}^{-1}$ for XG), particularly in the LPA and center of the circuit. More subtly, however, the high shear region on the outer curvature of each inlet limb was observed to extend further toward the opposite wall with XG compared to glycerol. This highlighted differences in flow separation points between the two solutions.

3.3 Shear Stress

As illustrated in Figure 7, the general shear stress distribution was similar between the solutions in the inlet and outlet limbs (mean for all conditions $1.2 \pm 4.7 \text{ dyn/cm}^2$ for glycerol vs. $1.3 \pm 4.7 \text{ dyn/cm}^2$ for XG). The magnitude of shear stress was higher with XG in high-stress regions (e.g. vessel walls). In contrast, differences in the distribution of shear stress in the center of the circuit did not follow a consistent pattern. With both solutions, shear stress was much lower in this region, however, the relative area of low shear stress varied depending on the experimental condition.

3.4 Pulmonary Blood Flow

Pulmonary blood flow distribution was expressed as the percentage of total flow going to the RPA. Under all conditions flow was statistically significantly different between glycerol and XG, as shown in Table 2 (mean $50 \pm 2.2\%$ for glycerol vs. $52.7 \pm 1.9\%$ for XG). This difference was more apparent under the low cardiac output conditions (mean 47.8% for glycerol vs. 53% for XG). Interestingly, the SVC/IVC flow ratio did not impact this distribution.

3.5 Power Loss

Table 3 compares the calculated power loss between the two solutions. Under all conditions, the percentage of power loss was statistically significantly higher with XG compared to glycerol (mean $10 \pm 2.9\%$ vs. $5.6 \pm 1.3\%$). Under most conditions, the difference was nearly two-fold. Although absolute power loss increased at high cardiac output, percentage power loss actually decreased at high cardiac output. As with pulmonary blood flow, SVC/IVC flow ratio had a minimal influence on power loss.

4. Discussion

The assumption of blood as a Newtonian fluid is common in computational and in vitro experimental studies of cardiovascular biofluid mechanics. As the circumstances under which non-Newtonian behavior is important largely depends on the shear rates in the region of interest, this assumption is likely reasonable in large vessel arterial blood flow, where

shear rates are high. In general, the non-Newtonian properties of blood are only apparent at shear rates less than 100 s^{-1} [3]. However, experimental and numerical models have shown that non-Newtonian behavior has a significant influence in the microvasculature [4] and in a variety of settings where low-shear flow is more prevalent such as in ventricular assist devices and aneurysms [5,6,28]. For example, a laser Doppler velocimetry analysis of flow in a pediatric ventricular assist device demonstrated that using a Newtonian fluid generally overestimated wall shear stress and therefore underestimated areas of low wall shear stress/thrombogenic potential compared to a non-Newtonian fluid [5]. A more recent computational fluid dynamics simulation comparing a Newtonian model with two non-Newtonian models in a similar ventricular assist device showed increased turbulent kinetic energy and a larger area of flow recirculation in the non-Newtonian models [6]. Interestingly, in contrast to the Bachmann study the predicted wall shear stress was similar between the Newtonian and non-Newtonian models; this is potentially explained by the larger size of the device in the latter study. Similarly, in computational fluid dynamics simulations of intracranial aneurysms, Newtonian models overestimated wall shear stress in areas of stasis or slowly recirculating vortices, which would effectively underestimate the likelihood of thrombus formation and aneurysm rupture [28,30].

In a recent clinical study, we demonstrated that non-Newtonian behavior is relevant in patients with univentricular heart defects - in this cohort who had undergone the Glenn and Fontan procedure, elevated low-shear blood viscosity correlated with decreased pulmonary blood flow [27]. However, to the best of our knowledge there have been no published investigations of the role of non-Newtonian behavior in experimental models of the Fontan circulation. Two frequently evaluated Fontan optimization parameters are hydraulic power loss and hepatic blood flow distribution, both of which could theoretically be affected by non-Newtonian behavior. Power loss has been shown to have a significant inverse correlation with exercise capacity even in asymptomatic patients with the Fontan circulation [20]. The importance of the latter parameter is less intuitive, but is related to pulmonary arteriovenous malformations, abnormal intrapulmonary shunts that allow blood to pass through the lungs without being oxygenated [31]. These malformations are suppressed by a liver-derived substance known as “hepatic factor” and hence they typically develop after patients undergo an SVC-pulmonary artery anastomosis (Glenn shunt), which excludes hepatic blood flow to the lungs [31]. Pulmonary arteriovenous malformations can regress after the Fontan procedure, however, maldistribution of hepatic blood flow can result in persistence or progression of the malformations in lung segments deprived of hepatic factor [32].

Based on computational fluid dynamics simulations using a Newtonian fluid assumption, two major revisions of the conventional tubular Fontan conduit were recently proposed: 1) the OptiFlo, which is comprised of two mirror image bifurcating limbs from each vena cava to both pulmonary arteries [25]; 2) the Y-graft Fontan, which is comprised of only a bifurcated graft from the IVC [19]. Of the two, only the Y-graft Fontan has been tested clinically. Furthermore, despite the promising performance metrics predicted by computational fluid dynamics, the clinical success of the Y-graft Fontan has been limited [19,26]. The largest clinical trial comparing a commercially available Y-graft to the conventional lateral tunnel and extracardiac conduit types of Fontan demonstrated increased

resistance, increased power loss in the setting of pulmonary artery stenosis, and similar or worse hepatic blood flow distribution with the Y-graft [26]. A much smaller study of custom-made Y-grafts showed that conduit performance was highly patient-specific [19]. Of note, one patient in this series developed complete occlusion of one limb of the Y-graft, which corresponded to a large area of low wall shear stress on retrospective computational fluid dynamics analysis [19]. While venous thrombosis is not uncommon in patients with the Fontan circulation, complete occlusion to this degree has rarely been reported with conventional extracardiac and lateral tunnel Fontan conduits [33-35].

Our data demonstrate that even in a simplified model of the Fontan circulation, significant differences exist between Newtonian and non-Newtonian fluids. Consistent with prior experimental studies of non-Newtonian behavior in vascular models, the velocity profile of the non-Newtonian fluid (XG) was blunted compared to that of the Newtonian fluid (glycerol) [36,37]. Qualitatively, flow patterns in the central portion of the circuit were also markedly different between glycerol and XG. While both fluids exhibited a large area of stagnation in this region under all conditions, they varied in terms of which conditions resulted in the formation of recirculations/eddies. In a patient with the Fontan circulation, eddy formation in this region would be expected to increase overall power loss and impede symmetric distribution of hepatic factor between the lungs.

Also consistent with other experimental evaluations of non-Newtonian behavior in vascular models [37,38], the distribution of shear stress was also significantly different between the two solutions. Wall shear stress was higher in the inlet and outlet limbs generally by 20-40% with XG compared to glycerol. The extension of the high shear region on the outer curvature of each inlet stream into the ipsilateral outlet stream was also more pronounced with XG. In contrast, in the central region of the circuit the distribution of low shear stress also differed, but did not demonstrate a consistent pattern between experimental conditions. Since the shear stress values observed in this study were all in the range of what would be considered abnormally low physiologically, it is difficult to make any inferences about how these differences might contribute to thrombus formation. Nevertheless, these differences do highlight the importance of non-Newtonian behavior in areas of low shear stress.

Most notably, power loss was consistently underestimated with glycerol vs. XG on average by 80%. This difference was most likely due to increased wall shear stress with XG, resulting in increased frictional energy loss. Eddy formation also appeared to contribute to power loss, but to a lesser degree than wall shear stress. For example, in both SVC/IVC 60/40 conditions, where eddies were more prominent with glycerol, power loss was only minimally higher compared to the respective SVC/IVC 50/50 conditions (6.8 vs. 6.6% for low cardiac output, 4.6 vs. 4.3% for high cardiac output).

Limitations of this study include the relatively simple circuit design in terms of vessel geometry, lack of elements outside of the region of interest, and lack of vessel compliance. Flow visualization was only performed in a single two-dimensional plane, so out of plane motion could not be characterized and mixing of the inlet streams could not be quantified. The relative coarseness of the mesh and inability of XG to demonstrate the subtle two-phase behavior of blood flowing through small blood vessels (i.e. a non-Newtonian red blood cell-

rich core surrounded by a Newtonian cell-free layer) also precluded precise calculation of wall shear stress at the vessel walls [4], although this was not necessarily significant since the primary goal of the study was to explore differences between the two fluids. Finally, the fact that pulmonary blood flow distribution was minimally affected by SVC/IVC flow ratio suggested that interactions in the common downstream connection of the outlet limbs negated the effects of the asymmetric inlet flow. This highlights the importance of downstream effects in experimental conditions and boundary conditions in numerical simulations [39,40].

5. Conclusion

In this simplified rigid wall model of the Fontan circulation, non-Newtonian behavior had a significant influence on overall flow patterns observed by particle image velocimetry. In particular, power loss and caval blood flow mixing, two frequently cited Fontan optimization parameters, were markedly different. In our study, power loss was consistently underestimated with a Newtonian fluid, predominantly due to decreased wall shear stress and different fluid flow behavior. Thus, assuming blood to be a Newtonian fluid introduces considerable error into computational models and in vitro hydraulic experiments of the Fontan circulation, where low-velocity low-shear flow predominates.

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Abbreviations

XG	xanthan gum
SVC	superior vena cava
IVC	inferior vena cava
RPA	right pulmonary artery
LPA	left pulmonary artery

Highlights

- A Newtonian fluid assumption is typically used in cardiovascular simulations.
- Non-Newtonian behavior has significant influence in regions of low-shear flow.
- We evaluated non-Newtonian behavior in a model of the low-shear Fontan circulation.
- Non-Newtonian behavior affected flow patterns and shear stress distribution.
- The Newtonian fluid consistently underestimated power loss vs. non-Newtonian fluid.

Fig. 1.

Viscosity measurements of 0.04% xanthan gum solution closely matched published viscosity data of Fontan patients. For interpolation between data points, viscosity was estimated with a Herschel-Bulkley model for shear rates of $0-30 s^{-1}$, a Casson model for $30-300 s^{-1}$, and as a constant Newtonian viscosity for $300-1000 s^{-1}$.

Fig. 2.

Schematic of the Fontan circuit. The superior and inferior cava (SVC and IVC) are the inlets. The right and left pulmonary arteries (RPA and LPA) are the outlets. Pressure (P) and flow (Q) were continuously recorded at each circuit inlet and outlet. Bold arrows indicate the direction of fluid flow.

Fig. 3.

Streamlines highlighted significant flow pattern differences in the center of the circuit. In particular, eddy formation was markedly different between xanthan gum (XG) and glycerol (Gly) in both SVC/IVC 60/40 conditions. All cases in the left column are under low cardiac output ($CO = 1.5$ L/min) and all cases in the right column are under high cardiac output ($CO = 2.5$ L/min). The top two rows are under a SVC/IVC distribution of 50/50 and the bottom two rows are under a SVC/IVC distribution of 60/40.

Fig. 4.

Xanthan gum (XG) and glycerol (Gly) demonstrated similar velocity distributions in the inlet and outlet limbs, however, the velocity profiles were more blunted with XG. In all conditions, there was a large area of stagnation in the center of the circuit. All cases in the left column are under low cardiac output (CO = 1.5 L/min) and all cases in the right column are under high cardiac output (CO = 2.5 L/min). The top two rows are under a SVC/IVC distribution of 50/50 and the bottom two rows are under a SVC/IVC distribution of 60/40.

Fig. 5. Xanthan gum (XG) and glycerol (Gly) demonstrated similar shear strain distributions in the inlet and outlet limbs, however, the average and maximum shear strain magnitude was greater with glycerol. The high shear region on the outer curvature of each inlet limb also extended further toward the opposite wall with XG, highlighting the differences in flow separation points between the two solutions. Insets show the center region at higher spatial resolution. All cases in the left column are under low cardiac output ($CO = 1.5$ L/min) and all cases in the right column are under high cardiac output ($CO = 2.5$ L/min). The top two

rows are under a SVC/IVC distribution of 50/50 and the bottom two rows are under a SVC/IVC distribution of 60/40.

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Fig. 6.

The magnitude of shear stress was higher with xanthan gum (XG) in high-stress regions (e.g. vessel walls). Differences in the distribution of shear stress in the center of the circuit did not follow a consistent pattern. Insets show the center region at higher spatial resolution. The relative area of low shear stress was variable - the graphs show percentage of the center area with low shear stress at different thresholds of 'low shear.' The top two rows are under a SVC/IVC distribution of 50/50 and the bottom two rows are under a SVC/IVC distribution of 60/40. The top row of each set is under low cardiac output (CO = 1.5 L/min) and the bottom row of each set is under high cardiac output (CO = 2.5 L/min).

Table 1
Comparison of Reynolds number between glycerol and xanthan gum

		Glycero	l Xanthan gum
SVC/IVC 50/50 Low CO	SVC	544	214
	IVC	712	224
	RPA	526	240
	LPA	536	127
	Center	81	3
SVC/IVC 50/50 High CO	SVC	864	390
	IVC	1043	391
	RPA	792	326
	LPA	744	211
	Center	89	22
SVC/IVC 60/40 Low CO	SVC	646	264
	IVC	612	189
	RPA	548	214
	LPA	533	158
	Center	4	6
SVC/IVC 60/40 High CO	SVC	1008	514
	IVC	910	357
	RPALPA	791821	322171
	Center	45	29

Table 2
Comparison of pulmonary blood flow (% flow to RPA) between glycerol and xanthan gum

	Glycerol	Xanthan gum	P
SVC/IVC 50/50 Low CO	47.8±1.3%	53.3±1.2%	<0.0001
SVC/IVC 50/50 High CO	52.1±0.9%	51.2±0.6%	<0.0001
SVC/IVC 60/40 Low CO	47.7±1.3%	52.6±1.2%	<0.0001
SVC/IVC 60/40 High CO	52.2±0.9%	53.5±0.7%	<0.0001

Table 3
Comparison of percent power loss between glycerol and xanthan gum

	Glycerol	Xanthan gum	P
SVC/TVC 50/50 Low CO	6.6±0.1%	12.3±0.1%	<0.0001
SVC/TVC 50/50 High CO	4.3±0.1%	9.5±0.1%	<0.0001
SVC/TVC 60/40 Low CO	6.8±0.01%	12.1±0.1%	<0.0001
SVC/TVC 60/40 High CO	4.62±0.1%	6.1±0.01%	<0.0001