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Computational Fluid Dynamics Analysis to Determine Shear Stresses and Rates in a Centrifugal Left Ventricular Assist Device

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Abstract

Axial flow left ventricular assist devices (LVADs) are a significant improvement in mechanical circulatory support. However, patients with these devices experience degradation of large von Willebrand factor (vWF) multimers, which is associated with bleeding and may be caused by high shear stresses within the LVAD. In this study, we used computational fluid mechanics to determine the wall shear stresses, shear rates, and residence times in a centrifugal LVAD and assess the impact on these variables caused by changing impeller speed and changing from a shrouded to a semi-open impeller. In both LVAD types, shear rates were well over 10 000/s in several regions. This is high enough to degrade vWF, but it is unclear if residence times, which were below 5 ms in high-shear regions, are long enough to allow vWF cleavage. Additionally, wall shear stresses were below the threshold stress of 10 Pa only in the outlet tube so it is feasible to endothelialize this region to enhance its biocompatibility.

Keywords

Left ventricular assist device; Hemodynamics; Shear; Fluid dynamics; Centrifugal; von Willebrand factor; Endothelium

Axial flow left ventricular assist devices (LVADs) significantly improve survivability for bridge-to-transplant patients and for destination therapy patients ineligible for transplant (1,2). However, centrifugal LVADs, which are currently under development, have the design advantage of a levitated impeller, which reduces wear, and are therefore considered the next generation in LVAD technology. This study investigates the fluid dynamics and, specifically, shear stress and shear rate of a centrifugal LVAD design based on, but not identical to, the HeartMate III (Thoratec, Pleasanton, CA, USA). The fluid dynamics of the HeartMate III were studied previously, but shear stresses and rates were not reported (3). The HeartMate III has a larger primary flow path than axial flow LVADs, allowing it to

produce adequate flow at lower impeller velocities. Therefore, the HeartMate III may produce lower wall shear stresses than axial flow devices.

The high-shear environment in LVADs may damage von Willebrand factor (vWF), which is responsible for platelet attachment to subendothelial layers under high shear stresses (4). vWF multimers begin to unfold at shear rates above 1000/s in saline, reaching lengths of 15 μm above a critical shear rate of 5000/s (5). Elongation of vWF exposes binding sites for platelets and subendothelial collagen, facilitating platelet attachment to damaged vessel walls. Because larger molecules are more prone to elongate in shear flows, high-molecular weight vWF multimers are primarily responsible for platelet-binding activity. However, vWF proteolysis by the enzyme ADAMTS13 increases under increased shear (6,7). Tsai et al. found a decrease in large vWF multimers in plasma at shear rates of 3351/s and higher over 12 s, which is, to our knowledge, the shortest residence time used in any vWF cleavage study (6). Fluid shear stress enhances cleavage of vWF because the unfolding of vWF multimers that occurs under high shear exposes a binding site for ADAMTS13 on the A2 domain of vWF (8). Zhang et al. demonstrated this by applying tension to A2 domains, resulting in unfolding at approximately 12pN. Unfolded A2 domains rarely ruptured without ADAMTS13 present, but ruptures increased exponentially over time in the presence of ADAMTS13.

vWF cleavage in shear flow increases with multimer length and number of platelets attached and is also most likely when the multimer is oriented at 45 degrees to the direction of flow, which results in the highest tension on the multimer (9,10). These and other constraints make mathematical determination of the threshold shear for vWF cleavage difficult. Therefore, the amount of time that the A2 domain of vWF needs to be exposed to ADAMTS13 to be cleaved in LVADs is currently unclear.

In spite of uncertainty about the critical shear rate and exposure time, vWF deficiency is a clinical problem in patients with axial flow LVADs, and multiple studies show that patients with these devices have low vWF activity (11–13). For example, vWF activity was reduced in HeartMate II (Thoratec, Pleasanton, CA) patients and that low vWF activity was associated with severe bleeding (11). Furthermore, studies indicate that patients with continuous flow LVADs have an increased risk of gastrointestinal (GI) bleeding (11,14–16). One study of over 100 LVAD patients found that those with continuous flow LVADs were nearly 10 times as likely to have GI bleeding as those with pulsatile LVADs (14). vWF deficiency is also associated with GI bleeding in patients with aortic valve stenosis (17). In these patients, vWF deficiency results from high shear stresses in the stenosed region.

While there is currently too little clinical data to know if acquired von Willebrand disease is a problem within centrifugal LVADs, quantifying the high-shear environment within LVADs may help determine if we can expect vWF damage and subsequent bleeding problems in patients. Therefore, this study uses computational fluid dynamics (CFD) to determine shear rates throughout a centrifugal LVAD and residence times in high-shear regions at flow rates of 4 and 7 L/min. Because the shrouds on the HeartMate III impeller are designed to produce high-shear back-flows, we also compare the shrouded impeller design to a semi-open design with no shroud on top of the impeller. Additionally, because

there is interest in lining the LVADs' inner walls with cells to improve biocompatibility, we determined the shear stresses on these walls to assess the feasibility of endothelialization (18,19).

MATERIALS AND METHODS

Geometry and mesh

The geometry and mesh of the LVAD were created using Gambit software (ANSYS, Inc., Canonsburg, PA, USA). The dimensions were derived from scaling previously published figures of the HeartMate III using as references the diameter and height of the device, 69 and 39 mm, respectively (2,20,21). The HeartMate III was chosen partly because its dimensions and clear figures were readily available. Rather than model the inlet tube leading to the main volute of the HeartMate III, blood flowed directly into the main volute with a constant velocity of 1.5 or 0.85 m/s for 7 or 4 L/min of flow, respectively, which was calculated based on flow rate and inlet area. Burgreen et al. reported that blood entering the impeller quickly disperses outward so constant inlet velocity should not have a major impact on the solution (3).

The structured, finite-volume mesh was composed of 1 062 296 hexahedral and pentahedral elements (see Fig. 1). Simulations with a denser mesh (1 250 000 mesh volumes) produced similar results. The structured mesh was used because it allowed complete control of density throughout the mesh, which was made densest nearest to the blade tips to accurately calculate large velocity gradients expected in that region. This method has been used previously in LVAD design and would not have been possible with an unstructured tetrahedral mesh (22). Furthermore, structured hexahedral elements were aligned with the direction of flow in areas of predictable flow, such as the outflow. This is recommended in Fluent (ANSYS, Inc.) to reduce error due to numerical diffusion and is not possible with tetrahedral elements. The same mesh was used for both shrouded and semi-open designs. The mesh faces used to define the top shroud of the impeller in the shrouded design was changed from a wall boundary to an interface boundary in the semi-open design.

Blood flow through the LVAD was simulated in Fluent 12.0.16 (ANSYS, Inc.). Due to high shear rates in the device, blood was modeled as a Newtonian, incompressible fluid with a density (ρ) of 1050 kg/m³ and viscosity (μ) of 0.0035 Pa/s. Isothermal conditions were assumed because the heat produced by the HeartMate III is quickly carried away by blood flow, resulting in a temperature increase of no greater than 1°C (21).

CFD and postprocessing setup

The impeller was set in motion using a sliding mesh model. The angular velocity of the volumes within the sliding mesh was set at 4500 rpm, which corresponds to the impeller speed used previously to generate 7 L/min of flow (3). An impeller speed of 3750 rpm was determined for the 4 L/min flow.

Preliminary simulations showed high Reynolds numbers (Re) near the impeller, indicating turbulent flow but lower Re elsewhere in the pump. Therefore, the shear stress transport (SST) k- ω model (23)—a blend of the k- ϵ and k- ω models that allows for use over a wide

range of Reynolds numbers—was used to simulate flow in much of the pump. In the outflow pipe, an area of low Re flow with a parabolic velocity profile was defined as a laminar flow region in order to reduce computational time. Reynolds number was estimated exiting the interblade region of the impeller with the formula $Re = 2\rho vwh/(\mu[w + h])$, where v is velocity estimated from Figs. 2 and 3 and w and h are blade height and distance, respectively, between outer blade tips. For the inlet, $Re = \rho vd/\mu$ was used, where d is inlet diameter.

While most of the data were presented using contour plots, path lines plotted from the inlet and colored by residence time were used to estimate residence time of particles in high-shear regions. For each condition, eight path lines that were spaced apart circumferentially such that no one section of the radial gap was overrepresented were averaged to find residence times. All data displayed here were collected after four impeller rotations. Simulations produced similar results after four and five impeller rotations, indicating that fluid dynamics in the pump were approaching a consistent, periodic state. After simulations were completed, figures were scaled and processed in Tecplot 360 (Tecplot, Inc., Bellevue, WA, USA).

RESULTS

Fluid speed under 4 or 7 L/min flow rates is displayed in Figs. 2 and 3 on a plane cut through the center of the LVAD and perpendicular to the outlet tube. These flow rates were chosen to provide a physiologically realistic range of flows and to compare to previously published results for the HeartMate III at 7 L/min. Velocity and shear rate are slightly higher in the shrouded impeller design and at 7 L/min (Figs. 2 and 3). Turbulence is also higher at 7 L/min as is shear stress, particularly above the semi-open impeller (Figs. 4 and 5). Turbulence was highest in the vicinity of the impeller, and Re estimates are found in Table 1. Given that inlet Re values were 2550 and 4500 for 4 L/min and 7 L/min of flow, respectively, our decision to use the SST $k-\omega$ model in anticipation of turbulence was justified. Table 1 indicates that blood flow increasingly transitions to turbulence in the impeller.

Pressure reached a maximum near 25 000 Pa in a small, outer region of the upper volute and exhibited a pattern similar to previous pressure plots of the HeartMate III, with high pressure in outer portions of the pump decreasing as blood moves toward the bottom and radially toward the center of the pump (data not shown). Shear rates through most of the central core and outer pump volume were much lower than those between impeller blades and along the walls of the shrouds, with the highest values of over 20 000/s for the 4 L/min flow and over 30 000/s for the 7 L/min flow at the blade tips and in the narrow radial gap between the lower shroud and side wall. Residence times of path lines through the radial blade clearance gaps, areas with very high shear rates, were below 5 ms and did not vary significantly between LVAD designs or flow rates (Table 1). Order of magnitude estimates using an average axial velocity (V) of 3 m/s and the formula $\text{displacement} = Vt$ gave a residence time t of 3 ms, similar to computed values reported in Table 1.

Shear stresses on the upper side wall and top of the LVAD are displayed in Figs. 4 and 5, respectively. Shear stresses were predominately below 10 Pa on the bottom of the outlet pipe and below 100 Pa on the bottom of the upper volute for both LVAD designs and flow rates (data not shown). On the lower side walls, shear stresses remained mainly above 100 Pa for all cases, below 150 Pa at 4 L/min, and below 200 Pa at 7 L/min (Table 1). Generally, shear stresses were higher with shrouded impellers than with semi-open impellers, with the highest wall shear stresses located on the top wall above the blades of the semi-open impeller.

DISCUSSION

While previous models of the HeartMate III assumed laminar flow (3), turbulence arose in the inner core of the lower shroud (data not shown), creating areas of lower velocity on its inner walls, especially with the semi-open impeller. If not endothelialized, these areas may be more prone to thrombus than previously thought; higher shear elsewhere on this inner wall may make it difficult to maintain an endothelium on the entire wall. Otherwise, the shrouded LVAD analyzed here showed many similarities to Burgreen et al.'s description of the HeartMate III, with similar flow speeds and pressure-driven, high-shear backflows in the radial gap and above and below the impeller shrouds (3). The maximum pressure of 25 000 Pa on the left side of the outer volute is higher than maximum pressure in the HeartMate III (18 000 Pa) and may be due to turbulence in the impeller region. As is evident in Figs. 2 and 3, the outer wall of the LVAD is closer to the impeller on the left side, where the highest pressures occur so it is possible that nearby turbulence between the impeller blades affected the maximum pressure in this study. It is notable, however, that pressures in the high-shear regions below and radial to the impeller blades are in the ranges of 6000–10 000 and 10 000–15 000 Pa, respectively, for both LVADs. Pressures in the current study did not change with increases in mesh density.

Comparison of shear stress values with other LVADs is difficult due to the scarcity of published shear data for other centrifugal LVADs, but an early study of the HeartQuest LVAD reports maximum shear stress of 427.7 and 446.4 Pa at 6 and 8 L/min of flow, respectively (24). In blood, these stresses convert to shear rates of over 120 000/s. This is significantly higher than shear rates in the current study, which are below 40 000/s in both designs, and residence times in high-shear regions was 20–25 ms in the HeartQuest, five times longer than the residence time in the high-shear radial gap of the LVAD studied here.

While we can consistently adhere a confluent endothelium to titanium at up to 10 Pa (18), most walls of the pump housing experience shear stresses well above this threshold, even at 4 L/min of flow. The exception to this is the short outlet pipe between the main section of the LVAD and the cylindrical outflow. The top, bottom, and right-hand side wall (Fig. 4) of this region remain almost entirely below 10 Pa of shear stress at 4 L/min and may therefore be good candidates for endothelialization. Shear rates in the high-shear backflows and near the impeller blades are high enough to unfold vWF multimers and expose the A2 domain, but residence time in the radial gap between the lower shroud and housing, which had the highest shear rate of any defined region, was below 5 ms. It is currently unclear if this is

enough time for ADAMTS13 to attach to and cleave the A2 domain of vWF, but these data will become more useful once the minimum time needed for vWF cleavage is determined.

In addition to limited knowledge of vWF cleavage, this study is limited by a lack of experimental results. A previous study of the HeartMate III reported that CFD largely agreed with experimental results, with minor variations of less than 10% (3). Also, due to computational expense, residence times were only calculated where shear rates were highest. Other regions of the pump may also have potential to damage vWF, and future studies should determine residence times throughout LVADs.

CONCLUSIONS

Based on wall shear stresses, three walls of the outlet tube are the only parts of the LVAD studied here that can be endothelialized. However, this may be insufficient to prevent thrombus in the pump. Similar to many LVADs, the HeartMate III is designed to have high-shear flows that prevent thrombus from building up on its walls. To fully endothelialize the pump housing, an LVAD designed specifically with endothelial lining in mind, with wider flow paths and lower shear stresses, would be necessary. Wider flow paths may reduce the hydrodynamic efficiency of the pump, and this cost would need to be weighed against the antithrombogenic benefits of endothelium. Additionally, shear rates near the impeller are high enough to result in vWF cleavage at both flow rates studied here, but short residence times in high-shear regions of the pump may prevent significant deficiencies in large vWF multimers.

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FIG. 1.
Finite volume mesh. (A) Top view of complete LVAD; (B) side view of complete LVAD.

FIG. 2. Velocity magnitude (m/s) on a planar cut through a centrifugal LVAD: (A) semi-open impeller, 7 L/min flow; (B) shrouded impeller, 7 L/min flow; (C) semi-open impeller, 4 L/min flow; (D) shrouded impeller, 4 L/min flow. The two dark gray rectangles in the lower volute are not fluid but rather cross-sections of the lower shroud.

FIG. 3.

Shear rate (1/s) on a planar cut through a centrifugal LVAD: (A) semi-open impeller, 7 L/min flow; (B) shrouded impeller, 7 L/min flow; (C) semi-open impeller, 4 L/min flow; (D) shrouded impeller, 4 L/min flow.

FIG. 4. Shear stress (Pa) on the upper side wall of a centrifugal LVAD: (A) semi-open impeller, 7 L/min flow; (B) shrouded impeller, 7 L/min flow; (C) semi-open impeller, 4 L/min flow; (D) shrouded impeller, 4 L/min flow.

FIG. 5. Shear stress (Pa) on the top wall of a centrifugal LVAD: (A) semi-open impeller, 7 L/min flow; (B) shrouded impeller, 7 L/min flow; (C) semi-open impeller, 4 L/min flow; (D) shrouded impeller, 4 L/min flow.

TABLE 1

Estimates of maximum Reynolds number between impeller blades and shear and residence times in radial gap between the lower shroud and wall

Impeller type, flow rate	Shrouded, 7 L/min	Semi-open, 7 L/min	Shrouded, 4 L/min	Semi-open, 4 L/min
Average residence time (standard deviation) (ms)	4.4 (0.32)	4.1 (0.47)	4.0 (0.39)	4.1 (0.14)
Reynolds number	14 590	11 670	8363	8363
Shear stress on lower side walls (Pa)	<200	<200	<150	<150