Fluid Dynamics of a Centrifugal Left Ventricular Assist Device

by

Brian P. Selgrade

Department of Biomedical Engineering
Duke University

Date:_______________________
Approved:

___________________________
Dr. George A. Truskey, Supervisor

___________________________
Dr. David F. Katz

___________________________
Dr. Edward J. Shaughnessy

Thesis submitted in partial fulfillment of
the requirements for the degree of Master of Science in the Department of
Biomedical Engineering in the Graduate School
of Duke University

2010
ABSTRACT

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Abstract

High shear stresses and shear rates in left ventricular assist devices (LVADs) make endothelialization of the LVAD difficult and likely contribute to cleavage of large von Willebrand factor multimers and resulting bleeding problems in patients. To better understand shear in a centrifugal LVAD, flow was simulated using finite volume and computational fluid dynamics (CFD) analysis. The $k$-$\omega$ model simulated turbulence and sliding meshes were used to model the movement of the impeller. CFD results showed high-shear backflows in the radial gap between the impeller and the volute wall, but residence times in this region were under 5ms. It is unclear if this is sufficient to cleave VWF, and more study is necessary to determine if other areas in the LVAD have potential for VWF cleavage. Although the walls near the outlet experience low shear stress and may be good candidates for endothelialization, shear stresses above 20-30Pa on all other walls of the pump make the possibility of endothelial cell growth elsewhere in the LVAD unlikely. An LVAD designed specifically to have low shear may be a better candidate for endothelialization.
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1. Introduction

Left ventricular assist devices (LVADs) are used in heart failure patients to help pump blood through the body. High shear stress in LVADs can result in blood damage, and thrombus on LVAD walls is sometimes a problem as well. Therefore, in this study, we analyzed shear stress and shear rate in a centrifugal LVAD to determine if conditions will encourage the cleavage of von Willebrand factor (VWF) or allow for endothelialization of the walls, which would reduce thrombus.

Heart failure results in about 300,000 deaths annually in the United States (1), but donor hearts are scarce, with the waiting list normally over 2,500 patients (2). Therefore, LVADs were developed to bridge patients to heart transplant, to provide destination therapy for transplant-ineligible patients and to unload the heart muscle so it can recover and the LVAD can be removed.

First generation LVADs were designed to mimic the heart’s pulsatility and include the HeartMate XVE and the Novacor Left Ventricular Assist System (3). These pulsatile LVADs are now less frequently used due to high rates of stroke in Novacor patients and mechanical problems in the HeartMate XVE. Second generation, or axial-flow, LVADs are much smaller than first generation devices and have magnetically driven impellers held axially between two bearings (4). A recent study comparing the HeartMate XVE to the HeartMate II – an axial-flow LVAD – shows that patients on the axial-flow device had higher survival and quality of life and less need for device replacement (5). Third generation LVADs
are magnetically driven centrifugal pumps with suspended impellers that do not contact other parts of the pump. Both second and third generation devices are designed with the impeller as the only moving part in order to reduce mechanical problems, and several models have inner coatings to reduce thromboemboli. Despite initial concerns over continuous flow, studies show no marked increase in adverse events in patients with continuous flow LVADs (6), (7). The primary goal of this study is to determine shear stress and shear rate in a third generation LVAD. Shear stress (τ) and shear rate (\( \dot{\gamma} \)) are related by the formula

\[
(1) \quad \tau = \mu \dot{\gamma}
\]

where \( \mu = 0.0035 \) Pa-s is the dynamic viscosity of blood. Shear stress and rate are particularly important parameters in blood flow, because low values promote the formation of thrombus and high values can damage red blood cells, platelets and von Willebrand factor (VWF).

1.1 Second Generation LVADs

1.1.1 The MicroMed DeBakey VAD

The DeBakey VAD was one of the first axial flow VADs. Its flow chamber has 3 parts: a flow straightener and diffuser acting as bearings for the impeller, which is based on the Archimedes screw (8) (9). The wing-shaped flow straightener prevents rotation of blood before it enters the impeller in order to reduce shear stress, and the diffuser axially redirects blood flow from its angular trajectory in the impeller. Like all second generation devices, the DeBakey VAD has a
magnetically driven impeller. Shear stresses as high as 71 Pa in the flow straightener region of an early model of the DeBakey were determined using computational fluid dynamics (CFD) (8). Flow through the DeBakey transitions to turbulence as the flow rate increases, with Reynolds numbers ranging from 1,255 to 5,270 for flow rates between 2.5L/min and 8.4L/min.

1.1.2 The Jarvik 2000

The Jarvik 2000 is similar to the DeBakey with an axial, magnetically driven impeller, diffuser and smooth, titanium blood-contacting surfaces (10). The impeller is supported axially by ceramic bearings and has no flow straightener. Also, the Jarvik 2000 has no inflow cannula, but attaches directly into the left ventricle for simpler implantation. For 46 patients implanted with the Jarvik 2000 for destination therapy, survival was 70% at one year and 52% at 2 years after operation with an average support time of 402 days (11). The Jarvik 2000 received CE Mark approval in 2005.

1.1.3 The Thoratec HeartMate II

The HeartMate II is comparable to the DeBakey, but the impeller has 3 blades, which are smaller at the downstream end than those of the DeBakey. A CFD study of the HeartMate II showed microvortices that wash the blades of the flow straightener and diffuser, preventing thrombus formation (12), (13). The manufacturer reports an estimated Reynolds number of 5,500 in the inflow cannula, indicating turbulent flow. Using the formula

$$\tau = \frac{\mu V L}{3}$$  (2)
where \( L \) is the distance between the impeller blade and the side of the housing and \( V \) is tangential velocity at the blade tip. This formula is valid for laminar flow and may underestimate shear stress in turbulent regions of the pump, but it is useful to determine a rough estimate of shear stress outside the impeller region of axial-flow LVADs. Using a rotational velocity of 9,000rpm to find \( V \), we estimated a maximum shear stress outside the impeller region of 13.8Pa. Using formula (1), this gave a shear rate of approximately \( 3,960 \text{s}^{-1} \), which may be enough to unfold von Willebrand factor (VWF) and allow for its cleavage (see Table 1). Shear rates in the inter-blade region of this pump are not known but may be higher than \( 3,960 \text{s}^{-1} \).

A multicenter trial of 133 bridge-to-transplant patients with the HeartMate II demonstrated 68% survival at one year and an average support time of 168 days (14). A more recent trial of 281 HeartMate II patients demonstrated similar results 18 months after implant: 79% of patients had received a transplant, were still on LVAD support or recovered and had the LVAD removed (15). Of those patients still on LVAD support at 18 months, survival was 72%. Complications included bleeding requiring surgery (31% of patients in older study and 26% in more recent study), ventricular arrhythmia (20% and 24%), respiratory failure (26% and 26%), and right heart failure (17% and 19%). The leading causes of death in these two clinical studies were sepsis and stroke, with ischemic stroke being more common than hemorrhagic stroke. However, in a recent study of 133 HeartMate II patients, hemorrhagic stroke caused the most deaths (9% of
patients) (5). Although bleeding events are common with axial-flow LVADs, the anticoagulation protocol was similar in the two studies, and the reason for this discrepancy is not entirely clear. The FDA approved the HeartMate II in 2008 (16).

1.2 Von Willebrand Factor

1.2.1 Experimental Studies

Although axial-flow devices are more durable and, due to their small size, more suitable for a wider range of patients than pulsatile LVADs, there is concern that these devices damage VWF. VWF is primarily responsible for platelet attachment to subendothelial layers under high shear stresses (17). Normally globular in shape, VWF multimers begin to unfold at shear rates above 1,000/s in saline, reaching lengths of 15μm above a critical shear rate of 5,000/s (18) (see Table 1). Elongation of VWF exposes A1 and A3 domains that are normally on the interior of globular VWF. These domains contain large numbers of binding sites for platelets and subendothelial collagen and facilitate platelet attachment to damaged vessel walls. The chance that multiple platelets will adhere to a VWF multimer increases exponentially with multimer length (19), so high molecular weight VWF multimers are primarily responsible for VWF’s platelet-binding activity.

However, studies have shown increased VWF proteolysis by the enzyme ADAMTS13 under increased shear stress (19), (20), (21). For example, Shim and colleagues applied shear stresses ranging from 10 to 50dyn/cm² to a mixture
containing ADAMTS13, platelets and VWF, and the most large multimer cleavage occurred between 10 and 20 dyn/cm$^2$ (19). The authors reasoned that cleavage was highest at these lower shear stresses because attached platelets increase the tension on VWF due to shear flow, and platelets may have had more difficulty binding to VWF at the highest shear stresses. In fact, when platelets were removed from the mixture, VWF response to shear was dramatically reduced, further indicating that platelet-binding increases tension and elongation of VWF under shear flow.

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 (22). Zhang and colleagues first demonstrated this by applying cyclic axial force to individual A2 domains using laser tweezers, resulting in unfolding that began between 5 and 15pN, depending on variations in loading
rate between 0.1 and 1000pN/s. The fully unfolded A2 domains rarely ruptured without ADAMTS13 present, but ruptures increased exponentially over time in the presence of ADAMTS13. The A2 domain occasionally exhibited a transient, partially unfolded intermediate state, but it is unclear if this state is cleaved by ADAMTS13. Baldauf and colleagues further demonstrated the importance of the A2 domain by comparing wild type VWF with VWF containing a mutated A2 domain (23). ADAMTS13 cleaved only the wild type VWF. Additionally, their molecular dynamics simulation of VWF under tension produced an intermediate, partly unfolded state similar to the one found by Zhang et al.

While Zhang et al. have demonstrated that the VWF cleavage site is exposed when the A2 domain experiences around 12pN of tension, determining the shear stress in blood flow that results in this tension poses a much more complicated problem. Shear stress is more likely to result in cleavage of VWF if it is attached to a vessel wall or platelets (19). Also, the tension in the A2 region depends on its location within the VWF multimer and multimer length. For A2 at the end of the multimer, tension is proportional to the number on monomers in the multimer, N, while tension in the center of the multimer is proportional to N^2 (24). Taking this into account, Zhang and colleagues estimated a tension of 100pN for a VWF monomer in the center of a rigid, extended 200-mer at a 45^o angle to 100dyn/cm^2 shear flow. However, VWF relaxes and extends while tumbling through shear flow, and a rigid model is not appropriate unless the multimer is fully extended. For a non-rigid multimer, Zhang estimated a maximum tension of 10pN in the
same 200-mer under the same conditions. This value does not seem to be consistent with VWF multimer behavior under experimental flows, in which large multimers have been cleaved at shear stresses as low as 40dyn/cm\(^2\) (20).

The orientation of VWF with respect to the shear flow also affects tension in VWF, with the maximum tension occurring when the multimer is at a 45\(^\circ\) angle to the shear flow and minimum tension occurring at 0\(^\circ\), 90\(^\circ\) and 180\(^\circ\) to shear flow (25). VWF molecules tumble due to the vorticity of shear flow, raising questions about how long VWF multimers are exposed to tensions high enough to uncover the A2 domain. In flow with a shear rate of 600/s and viscosity of 0.01dyn·s/cm\(^2\), Shankaran and Neelamegham determined a rotational period of 66ms for VWF dimers by modeling them as dumbbells, but the period for a high molecular weight VWF multimer is likely longer due to its size. Zhang et al. found a force loading time of 0.4sec during each rotation, but this estimate is again based on a rigid VWF multimer (24). Notably, both Zhang et al. and Shankaran and Neelamegham assumed Stokes flow, which is intended for capillaries and other regions where viscous forces dominate and is therefore probably unsuitable for faster flows such as those in LVADs. It is also currently unclear how long the A2 domain of VWF needs to be exposed for ADAMTS13 to bind and cleave it.

**1.2.2 Von Willebrand Factor in Clinical Studies**

While experimental studies have not determined the exact threshold shear stresses and residence times that result in VWF cleavage in LVADs, clinical studies provide more consistent results. Multiple studies show that patients with
axial flow LVADs have low VWF activity (26), (27), (28). Geisen and colleagues found that patients with either the HeartMate II or a continuous-flow, biventricular assist device had significantly lower VWF activity than a control group of heart transplant patients (27). Kloviate et al. found that VWF activity was lower in HeartMate II patients and that low VWF activity was associated with severe bleeding (26). Several other studies indicate that patients with continuous flow LVADs have an increased risk of gastrointestinal (GI) bleeding (26), (29), (30), (31). 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 (29). VWF deficiency is associated with GI bleeding in patients with aortic valve stenosis (32). In these patients, VWF deficiency results from high shear stresses in the stenosed region, and GI bleeding ceases when the stenosed valve is replaced.

Letsou and colleagues did not suggest VWF deficiency as a possible cause of GI bleeding when they first discovered GI bleeding in 3 of 21 Jarvik 2000 patients (30). Rather, they hypothesized that lowered pulse pressure or increased intraluminal pressure in both LVAD patients and patients with aortic valve stenosis led to development of GI angiodysplasias, leading to GI bleeding. However, it seems unlikely that angiodysplasia, which often does not result in bleeding (33), would be the sole cause of GI bleeding, and VWF deficiency probably plays a role as well. Angiodysplasias create high shear stresses, for which large VWF multimers are necessary for optimal hemostasis (33). Among
patients with GI angiodysplasias, the ones with GI bleeding have significantly fewer high molecular weight VWF multimers, suggesting that a lack of these multimers likely contributes to bleeding from angiodysplasias. Thus, while large VWF multimer deficiency has not been proven with complete certainty as the cause of GI bleeding, mounting evidence indicates that the two are linked both in patients with aortic stenosis and continuous-flow ventricular assist devices (33), (26).

1.3 Third Generation LVADs

All third generation LVADs are centrifugal pumps, meaning that blood flows through an inlet at the top of the pump down into the center of the impeller. The impeller pushes fluid away from the center of the pump and toward the outlet. Third generation LVAD impellers are magnetically-driven and suspended by either magnets, fluid forces or both so that they make no contact with other parts of the pump. This gives them the advantage of reduced heat production and greater durability, making them more useful for long-term applications such as destination therapy. Additionally, most third generation models tend to be larger than second generation devices and can thus produce adequate flow rates at lower impeller speed. This could cause lower shear stresses and shear rates and may, therefore, result in less VWF damage than is seen with second generation devices. However, their larger size makes third generation LVADs less suitable for implantation in adolescents than axial-flow devices. Third generation pumps
are more diverse in their designs than axial-flow pumps and include the VentrAssist, DuraHeart, HeartWare, Levacor and HearMate III.

1.3.1 The Ventracor VentrAssist
The VentrAssist's low impeller speed – only 3,000rpm provides up to 10L/min of flow – is intended to reduce shear stress, and hemolysis and platelet damage caused by this device are low both in vivo and in vitro (34), (35). Regions of high shear above and below the impeller blades cause little hemolysis because residence time in these areas is only about 5ms. The VentrAssist is driven by magnetically, but, unlike other third generation LVADs, the impeller is suspended only by fluid forces (36), (35). Also, flow deflecting perpendicularly from its conical housing produces a radial restoring force on the impeller, preventing it from contacting the sides of the housing. The VentrAssist impeller is open, meaning that it has no shroud either above or below the impeller blades and thus avoids the high shear layer present with shrouded impellers (35). CFD analysis indicates a turbulent main flow, but flow in the bearings is laminar (37). Flow is primarily radial below the impeller but circumferential above the impeller.

1.3.2 The Terumo DuraHeart
Unlike the VentrAssist, the DuraHeart has 18 blades and a shrouded impeller that is magnetically suspended, with a secondary hydrodynamic bearing in case magnetic levitation fails (38), (39). However, the DuraHeart still has low enough impeller speed and shear stress that it causes little hemolysis or platelet damage (40). In animal studies without anticoagulation, the DuraHeart had few emboli.
and little thrombus. However, clinical studies of the DuraHeart still use anticoagulation (38).

Of 33 patients in the initial clinical trial in 4 European centers from 2004 to 2007, 27 either received heart transplants or were awaiting transplant 3 months after the DuraHeart implantation for a survival rate of 82% (39). As of June 2007, the average support time for DuraHeart patients was 338 days. The most common cause of death was cerebrovascular accident, which was likely caused by excessive anticoagulation and killed 4 of the first 11 patients. The remaining patients received reduced anticoagulation and none had cerebrovascular accidents. Other common adverse events included bleeding requiring surgery (12%) and right heart failure (27%). Although bleeding was much less common and right heart failure was more common in patients with the DuraHeart than with the HeartMate II, these comparisons should be viewed cautiously because of the small sample size in the DuraHeart study and differences in anticoagulation between the two studies. After the trial, 35 more patients received the DuraHeart, and, for all 68 patients, survival was 81% at 6 months and 77% at one year. The DuraHeart has received CE Mark approval and is in clinical trials in the United States (41).

1.3.4 The HeartWare Left Ventricular Assist System

The HeartWare is the smallest 3rd generation LVAD (42). Its inflow cannula is built into the device, allowing for simpler attachment to a beating heart. Its shrouded impeller has 4 blades and is suspended axially by a thrust bearing and
radially by magnets. In 90-day animal studies, levels of platelet damage and hemolysis were normal, and, although no sheep received anticoagulation, there was no evidence of thrombus in the LVADs. A study of 43 HeartWare patients in 5 centers was recently reported (43). In the first 180 days, 12% of patients died while 19% received heart transplants, and 7% recovered and had the device explanted. The average support time was 247 days, with 63% of patients awaiting transplants when the study concluded.

1.3.5 The WorldHeart Levacor VAD
The Levacor is the most recently developed third generation LVAD. Its impeller is suspended magnetically and semi-open, meaning that it has a shroud below it but is open at the top of the impeller (44), (45). Large clearances between the impeller and the housing are designed to reduce shear and subsequent blood damage. Using a semi-open impeller with larger clearances rather than a shrouded impeller reduces hemolysis (46). The Levacor’s blade tip clearances of 50 µm also reduce hemolysis, possibly because fewer red blood cells can enter the clearance (47).

Reynolds numbers of about 100,000 in the Levacor indicate turbulence (46). One concern is that turbulence causes more hemolysis than laminar flow at equivalent shear stresses (48). However, the Levacor design minimizes shear stress, and, in animal studies, calves demonstrated low hemolysis (45). For shear stresses at or below 200Pa, hemolysis increased only slightly for turbulent flow, so
turbulence should not increase hemolysis in the Levacor if shear stresses remain in this range (48). Clinical trials for the Levacor are currently underway (49).

1.3.6 The Thoratec HeartMate III

The impeller of the HeartMate III is shrouded and is partly located in a cavity in the bottom of the fluid chamber (50). It consists of five backswept blades on top of a thick washer-shaped lower shroud. Blood flows into the impeller through a hole in the center of the upper shroud directly above the hole through the lower shroud. All blood-contacting surfaces except the impeller are sintered to encourage growth of a pseudo-intima (51). The impeller is driven and levitated magnetically.

CFD and flow visualization studies and showed high-speed, high-shear secondary flows moving radially inward above and below the impeller (52). These backflows wash blood-contacting surfaces, preventing thrombus formation. The lower backflow also moves up the inner walls of the washer-shaped section, preventing thrombus in this area. Despite these high-shear backflows, animal tests demonstrated low hemolysis (51).

1.4 Relevance of Proposed Research:

Recently, patients with continuous-flow LVADs have been found deficient in large VWF multimers (27), and axial-flow device patients have increased rates of GI bleeding, which is associated with lack of large VWF multimers (32). There is a strong possibility that excessive shear within the LVAD causes this large VWF deficiency, but, because these discoveries are relatively recent, most published
CFD studies of LVADs focus on effects of shear on hemolysis and platelet damage and do not consider VWF. In some cases, neither maximum shear stresses nor shear rates are reported and results demonstrating low hemolysis and platelet damage are deemed sufficient (52), (51), (40). However, if possible, it would be best to prevent clinical problems such as large VWF multimer deficiency before devices become widely used clinically. We have this opportunity with third-generation devices, which are mostly still in animal testing or clinical trials. A more exact understanding of the effects of shear on VWF, combined with mapping of shear stresses, shear rates and residence times inside third-generation LVADs, will help us determine if VWF deficiency will be a problem in third generation LVADs. The goal of this study is to use CFD to provide information on shear and residence times in a third generation device based on the HeartMate III.

The shroud in the HeartMate III is designed to produce a high-shear backflow above the impeller, and research on a different centrifugal pump showed that switching from a shrouded to semi-open impeller with larger clearance above the impeller reduced hemolysis (46). Therefore, we will test shrouded models of our LVAD and compare the shear stresses and rates to those of LVADs that have semi-open impellers but are otherwise identical.

Another issue of concern is thrombus forming on the inner walls of LVADs. Thrombus can break off of these walls and clot other areas, which can cause ischemic attacks. Transient ischemic attack afflicted 15% of DuraHeart patients
and 5-6% of HeartMate II patients suffered ischemic stroke (39), (15), (14). To prevent thrombus accumulation, the inner surfaces of the HeartMate III are sintered with titanium microspheres, which encourage the growth of a native lining of “pseudoneointima” over these surfaces, reducing the risk of thromboembolism (50). However, the pulsatile HeartMate I has the same sintered surfaces, and analysis of the pseudoneointima in two of these pumps revealed that it was largely collagenous and that most of the cells in the lining were macrophages. A confluent lining of endothelial cells between the blood and titanium LVAD surfaces would be ideal to prevent thrombus in the pump. One of our colleagues at Duke Medical Center is currently studying adherence of endothelial cells onto titanium tubes under flow with the eventual goal of endothelializing the inner walls of an LVAD. In this study, we will also show whether or not wall shear stresses are too high in a centrifugal LVAD to allow endothelial cells to adhere.
2 Materials and Methods:

2.1 Scaling the LVAD:

Because the specific dimensions of the HeartMate III were not available and the company declined to provide them, published figures of the HeartMate III were scaled to provide the dimensions used for the centrifugal LVAD in this study. Figure 1B from reference 50 was scaled using the pumps published diameter of 69mm and height of 30mm in order to determine vertical and horizontal lengths in the flow chamber. Figure 7 from reference 52 was scaled to better determine the radial dimensions, and the curved blades in Figure 5 from reference 52 were scaled such that the blade tips had the same radial distance from the center as in Figure 1B. Eight points on the blade were scaled in order to fit a NURBS curve to them to create the radial edge of each blade. The narrow gap distances outside and below the impeller were too small to clearly distinguish, and their lengths were first estimated based on the scaled figures. After an initial CFD model was created, these dimensions were modified until the CFD model produced velocities similar to those in Figure 3B from reference 52.

Modeling the inlet tube leading to the main volute of the HeartMate III was not attempted here because there are no published figures that allow accurate estimations if its dimensions. Rather, the blood flowing directly into the main volute from where the inlet tube would have attached was given a constant velocity of 1.5m/s, based on figure 3b from reference 52. A constant velocity is not accurate here, but Burgreen and colleagues report that blood entering the
impeller quickly disperses outward because the impeller speed is much greater than the inlet velocity, so we do not expect the constant velocity to have a major impact on the solution.

2.2 LVAD Geometry:
All LVAD geometry and meshing was created using Gambit, a graphical user interface program created by Fluent. The scaled dimensions from discussed above were plotted as points. In general, points were connected to create edges, edges were connected to create faces, and faces were connected to create volumes (see Figure 2). Faces were created not only for the walls, inlet, outlet and parts of the impeller, but also for interfaces between the rotating impeller region and the rest of the pump. Additional faces were used to separate volumes that needed to be meshed individually to achieve more accurate results.

2.3 Mesh Creation:
Mesh creation followed the same pattern as the geometry, with construction of mesh edges followed by mesh faces and, lastly, mesh volumes. Each edge in the xy-plane was meshed with an appropriate number of intervals and grading based on the local geometry. Edges were graded to increase the density of intervals as they approached walls, blade tips or other obstacles in the fluid, because these regions have the steepest velocity gradients. The vertical mesh edges in the LVAD were all given uniform mesh density such that each interval along the edges was 0.25mm long. This mesh density was necessary throughout because there were many changes in the vertical geometry of the LVAD.
Next, mesh edges were used to create mesh faces using either a paved, tri-primitive or mapped meshes (Figure 1). Mapped mesh faces are appropriate for rectangles or similar shapes (Figure 1A). They are essentially Cartesian grids whose meshes are entirely determined by the surrounding mesh edges, giving the user complete control over the mesh. For example, a rectangular mapped mesh face bordered by two mesh edges with $x$ intervals each and two mesh edges with $y$ intervals each will have $xy$ number of elements, and the grading of the face will mirror that of the edges. Tri-primitive meshes allow meshing of triangular faces by separating the triangle into 3 quadrilateral mapped meshes in the same mesh face (Figure 1B). In paved mesh faces, which are useful for circles and more awkward geometries, the elements are automatically generated throughout the face by Gambit (Figure 1C). Paved mesh faces can be directed to create either triangular or quadrilateral mesh elements, and the number of intervals on mesh edges governs the number of edge-adjacent mesh face elements, providing the user with some control over paved meshes. However, mapped meshes used wherever possible in this study, because they allow for much more control over mesh density throughout the face.
Figure 1: Mesh Faces: Two Examples of Mapped Meshes (A), a Tri-primitive Mesh (B), and Meshes Paved with Triangular and Quadrilateral Elements (C)

To begin meshing this particular LVAD, a small, circular face in the center of the inlet was paved using quadrilateral elements (Figure 2B). Around this section, the remaining faces of the inlet were mapped with the mesh graded densest near the wall of the inlet. All other faces in the impeller region were also mapped, with circumferential mesh edges densest near the blades and radial mesh edges densest near the blade tips. The inter-blade fluid volumes were simple enough that they could be map-meshed in a process analogous to mapping mesh faces.
However, because the inlet contained a paved face, the volume below the inlet, which included the fluid in the central hole of the LVAD, had to be meshed using the Cooper tool. Gambit does not allow mapped meshes of volumes bordered by paved faces, but the Cooper tool can mesh these volumes by projecting the paved face down throughout the volume. Even with this tool, only “source faces” on the top and bottom of the volume may be paved, and side faces of the volume still have to have mesh maps. The other mesh volumes directly bordering the impeller were also paved (Figure 2B and 2C). Using a mapped mesh for the large, circular region outside of the impeller would have necessitated the same number of mesh intervals on its inner and outer circumferential edges. This would either result in a less dense, less accurate mesh along the outer part of the pump or an extremely dense, computationally expensive mesh in the impeller region. Therefore, the faces of this region in the xy-plane were paved with triangular elements, and the volume was meshed using the Cooper tool (Figure 2D). The crescent-shaped region between the main volute and outlet was initially meshed the same way. However, this created highly skewed elements in the upstream corner of the face. Highly skewed elements increase numerical diffusion, a source of error in CFD calculations, and prevent Fluent programs from running. Therefore, this corner was replaced with a separate, tiny, triangular face that could be successfully meshed with a tri-primitive scheme. The remaining face on the crescent-shaped volume was paved with triangular elements (Figure 2D and 2E).
A curving, half-cylinder volume was created to model the region adjacent to the curved outer wall of the LVAD (Figure 3A). Rather than paving the semi-circular cross-section of this volume, the cross-section was split into a triangle with a tri-primitive mesh and two mapped quadrilaterals with radial edges (Figure 3B). This allowed for more control when increasing mesh density near the wall. An outflow pipe, including a cylinder and transitional region connecting this cylinder to the original outlet face, was added even though it is not part of the actual LVAD. However, adding this long pipe allowed the outlet at the end to be modeled as an outflow boundary condition without knowing any of the outlet parameters. Both outflow volumes were meshed by projecting the meshes on the original outlet faces throughout the volumes using the Cooper tool (Figure 2F).

The separate volumes shown in Figures 2 and 3 were only separated for meshing purposes, and fluid is not restricted from moving between volumes. Other than the inter-blade volumes, all fluid volumes were defined as one continuum entity, while the blades and washer-shaped impeller section were defined as separate solid entities. The inter-blade volumes were defined as separate fluid entities only because they are part of the sliding mesh created to model impeller rotation (see sliding mesh section). The combined volumes of the LVAD have a total of 1,053,493 nodes and 1,062,296 mesh volumes (see Figure 4). Simulations with a denser mesh (approximately 1,250,000 mesh volumes) produced very similar results, indicating that the 1,062,296-volume mesh is sufficiently dense.
Figure 2: Meshes of Segment Volumes of LVAD. A) Impeller without Shroud; B) 2 Volumes: Fluid through Central Core and below Impeller, and Fluid above Impeller; C) Fluid in Radial Gap between Blades and Wall of Lower Volute; D) 4 Upper Volute Volumes: 2 Volumes in the Washer-shaped Region, 1 Crescent-shaped Volume, and 1 Rectangular Volume Nearest to the Outlet; E) Closer View of Crescent and Rectangular Volumes; F) Outflow Volume
Figure 3: A) Outer Section of Upper Volute; B) Cross-section of Outer Section of Upper Volute
2.4 CFD Setup:

The separate mesh volumes created in Gambit were uploaded into FLUENT version 12.0.16 and fused together to create the complete LVAD mesh (Figure 4). Blood flowing through the LVAD was modeled as an incompressible fluid with a constant density of 1050kg/m$^3$ and viscosity of 0.035dyn/cm$^2$. Here, blood can be treated as a Newtonian fluid with constant viscosity due to the high shear rates in the LVAD. Isothermal conditions were assumed, because the heat produced by the electronics in the HeartMate III is quickly carried away by blood.
flow, resulting in a temperature increase of only 1°C in the hottest part of the pump (53).

**2.4.1 Sliding Mesh Model:**

Next, the impeller was set in motion. The sliding mesh model is designed to be used for impellers located close to stationary walls or stators that interact with the impeller region, resulting in unsteady flow. Therefore, we used a sliding mesh to simulate impeller rotation here due to the proximity of the impeller to the wall, even though this was more computationally expensive than other options. Also, a previous CFD study of the HeartMate III used sliding meshes (52). The angular velocity of the volumes within the sliding mesh was set at 471.24rad/s, which corresponds to the impeller speed of 4,500rpm used in the previous CFD study to generate 7L/min of flow (52).

In order to apply the sliding mesh, mesh interfaces were created around the impeller. These were constructed from faces that had already been created and defined as interfaces in Gambit. Inner and outer interfaces were defined on the same coordinates, with inner interfaces defining the impeller volumes and inter-blade volumes, and outer interfaces defining the outer volumes. Each mesh interface was created by combining an outer interface with its corresponding inner interfaces. Mesh interfaces were created at the top, bottom and outside edges of the impeller and around the inner cylinder between the impeller and the hole through its center. For shrouded versions of the LVAD, the top mesh
interface was defined as a coupled wall interface. This automatically placed a wall on top of the impeller blades between the inner and outer interfaces. To ensure that the shroud moved with the blades, these newly created walls were set to the same angular velocity as the sliding mesh.

2.4.2 Turbulence – the k-ω Model:

Preliminary simulations with laminar flow showed that high Reynolds numbers were present near the impeller, indicating turbulent flow. Fluent provides several options for dealing with turbulence including the popular k-ε model, which is useful for fully turbulent flows, and the k-ω model, which is intended for turbulent flows with lower Reynolds numbers. Turbulence was relatively low in this LVAD with some adjacent areas of laminar flow as well, so we considered two modifications of these models: the renormalization group theory (RNG) k-ε model, and the shear stress transport (SST) k-ω model.

The RNG model makes corrections to the k and ε equations, allowing for more accurate low Reynolds-number calculations. The SST k-ω model blends the k-ε and k-ω models, allowing for use over a wide range of Reynolds numbers. Both the k-ε and SST models have been used in previous fluid dynamics studies of centrifugal LVADs (45), (54), (55), but the SST k-ω model was chosen over the RNG k-ε model in this case because it allowed for shorter computational time (see Appendix A for further discussion of turbulence models). A preliminary simulation of laminar pipe flow with the SST k-ω model indicates that this model
accurately predicts laminar flow as well. Finally, because Reynolds numbers were low in the outflow, it was defined as a laminar flow region to reduce computational time.

### 2.4.3 Solution Methods:

For pressure-velocity coupling, the Pressure-Implicit with Splitting of Operators (PISO) algorithm was used. PISO is recommended in the Fluent User Guide for all transient calculations as it allows for the quickest convergence in these cases. Both the momentum equation and turbulence parameters were solved using second-order upwind discretization. Residuals for velocity, continuity, and turbulence parameters were required to converge below an absolute criterion of 0.01 before the solver could proceed to the next time step. Second-order discretization takes longer to converge than first-order discretization but is recommended by the User Guide to reduce numerical discretization error whenever flow is not fully aligned with the mesh. Because the flow pattern was complicated and parts of this LVAD were meshed with triangular elements, it was not possible to create a mesh that was always aligned with the flow, so second-order discretization was necessary.

Because interactions between the rotating impeller and stationary pump housing result in unsteady flow, a transient solver was used with a time step of $5.333 \times 10^{-6}$ s. One full rotation of the impeller was equal to 2,500 time steps. Using larger time steps caused the solution to diverge. During the computation, 5 to 10
iterations per time step generally indicate an appropriate time step size, with many more iterations indicating that the time step is too large. For most computations in this study, around 40 iterations/time step were needed for residuals to converge for the first few time steps. However, as the calculation continued, this quickly shifted to the point where convergence occurred in the vicinity of 10 iterations for most time steps. Therefore, it was determined that a smaller time step, which would have potentially increased the computational expense of an already large simulation, was unnecessary.

### 2.5 Post-Processing/Gathering Data:

Vector plots were produced to show the direction of flow in different sections of the pump and to look for swirling flow. Contour plots displayed most of the relevant variables in the fluid, including pressure, velocity and wall shear stress. However, contour plots allowed for determining shear along the walls but not throughout the fluid. Therefore, pathlines, which were colored by shear rate, were plotted from the inlet and blade. From the data on these plots, shear stresses can be calculated with equation (1). Subsequently, pathlines were colored by residence time in the pump. This allowed for estimation of residence time of particles such as VWF molecules in high-shear regions by subtracting the time that the pathline entered the region from the time it exited the region. For clearer viewing, most figures were produced by displaying contours or vectors on planes.
cutting through the main volute of the LVAD. All data shown here was collected after 4 complete rotations of the impeller.

3 Results:
Contour plots of velocity and pressure are presented here to characterize flow through the LVAD and to compare it to the data in reference 52 for the HeartMate III device after which this LVAD was modeled. For this same reason, the flow rate was chosen to be 7L/min, the same as was used in reference 52. Because turbulence has been shown to increase blood damage, turbulent kinetic energy plots are presented here as well (48). I have included plots of wall shear stress to determine if conditions are appropriate for endothelialization of these walls. However, these plots give no information about shear throughout the fluid, so several contour plots of shear rate on different cross-sections of the LVAD are also included.

3.1 Velocity and Pressure
In Figures 5, 6 and 7, contour plots of speed, axial velocity and radial velocity are displayed on a plane through the center of the LVAD and perpendicular to the outlet tube. With the shrouded impeller, a region of high speed velocity moves up the inner wall of the lower impeller shroud, while the semi-open design creates small, downward fluid velocities along much of this wall. This upward flow exits the core of the impeller and quickly moves radially outward between the impeller blades (Figure 8). Radial velocity is generally directed toward the center above
and below the impeller in both designs (Figure 7). These backflows are driven inward by high static pressure in the outer part of the volute (Figure 9), which agrees with previous results for the HeartMate III (52). Velocity is primarily tangential to impeller motion in a mid-plane through the center of the impeller blades (Figure 10). Above the semi-open impeller, tangential velocity largely follows the pattern of the blades, whereas velocity is more evenly distributed above the shrouded impeller (Figure 11).
Figure 5: Fluid Velocity Magnitude (m/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure 6: Axial Velocity (m/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure 7: Radial Velocity (m/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure 8: Radial Velocity on a Transverse Mid-plane Cut through the Vertical Center of the Impeller Blades (m/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure 9: Static Gauge Pressure (Pa) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller. Inlet Pressure Set at Zero
Figure 10: Tangential Velocity on a Transverse Mid-plane Cut through the Center of the Impeller Blades (m/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure 11: Tangential Velocity on a Transverse Mid-plane Halfway between the Impeller and Top of the Volute (m/s). A) Semi-open Impeller; B) Shrouded Impeller
3.2 Turbulence

Turbulence is highest in the inter-blade regions and inner core of both impellers with little turbulence below the impeller and in the outer volute (figure 12). While the shrouded impeller has slightly higher turbulence in the inter-blade region, turbulence is slightly higher above the semi-open impeller (figure 13). Vector plots were used to locate eddies in the inner core of each impeller (figure 14).

3.3 Shear Rate, Wall Shear Stresses and Residence Times

Shear stresses on the walls of both LVAD designs are displayed in figures 15 through 19, and shear rates throughout the LVAD are shown on figures 20 through 23. Average residence times of path lines through the radial blade clearance gaps – areas with very high shear rates – are presented in table 2. Rough estimates using an axial velocity (V) of 3m/s and the formula

(3) \[ \text{displacement} = Vt \]

gave a residence time t of 3 ms. Formulas 1 and 2, the impeller velocity and the radial gap width were used to estimate a shear stress of around 40Pa and shear rates above 10,000/s. These values are in the vicinity of computational results.

Table 2: Residence Times in Radial Gap between Impeller and Wall after Three Impeller Rotations

<table>
<thead>
<tr>
<th></th>
<th>LVAD with semi-open impeller (n=8 pathlines)</th>
<th>LVAD with Shrouded Impeller (n=8 pathlines)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average Residence Time (st.dev.) in ms</td>
<td>3.2 (±0.13)</td>
<td>(3.8 ± 0.35)</td>
</tr>
</tbody>
</table>
Figure 12: Turbulent Kinetic Energy (m$^2$/s$^2$) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure 13: Turbulent Kinetic Energy on a Transverse Mid-plane Halfway between the Impeller and the Top of the Volute (m$^2$/s$^2$). A) Semi-open Impeller; B) Shrouded Impeller
Figure 14: Velocity Vectors (m/s) Colored by Axial Velocity with Eddies Circled on Vertical Plane through the Center of the LVAD. A) Semi-open Impeller; B) Shrouded Impeller
Figure 15: Shear Stress on Side Wall of Lower Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure 16: Shear Stress on Side Wall of Upper Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure 17: Shear Stress on Bottom Wall of Lower Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure 18: Shear Stress on Bottom Wall of Upper Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure 19: Shear Stress on Top Wall of Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure 20: Shear Rate on a Transverse Mid-plane Cut through the Center of the Impeller Blades (1/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure 21: Shear Rate on a Transverse Plane Midway between the Impeller and the Top of the Volute (1/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure 22: Shear Rate (1/s) on Vertical Plane Cut through the Center of LVAD with
A) Semi-open Impeller; B) Shrouded Impeller
Figure 23: Magnified View of Shear Rates in Radial Blade Clearance Gaps (1/s). A) Semi-open Impeller; B) Shrouded Impeller
4 Discussion:

Figures 15 through 23 show high shear stresses and shear rates in parts of both LVAD designs. Except for the walls closest to the outlet, wall shear stresses were above 50Pa and reached well over 300Pa on the top walls of both designs, with the highest shear rates on the top wall of the semi-open design. Shear rates reached above 10,000/s in several regions near the impeller and was at its highest in the radial gap between the impeller and lower volute wall, where residence times were less than 5ms. Although previous studies of the HeartMate III did not report exact shear stresses, this data is consistent with Burgreen et al.’s description of high shear backflows (52).

Velocity and static pressure in the shrouded impeller model studied here were slightly greater than in the HeartMate III but displayed similar patterns overall, with the highest pressures in outer parts of the pump and low flow velocity in the central core of the impeller.

Another interesting result is the amount of turbulence and swirling flow in the vicinity of the semi-open impeller (Figures 12 and 14). The only LVAD under development with a similar semi-open impeller is the Levacor, which has a more organized flow with no areas of recirculation (44). These differences are due to the different impeller and housing shapes between the two LVADs, and the results here should not be taken to mean that semi-open impellers cause more turbulence in general. Furthermore, the semi-open impeller actually resulted in
higher shear stress on the top wall than the shrouded impeller. Generally, an open impeller design results in less shear than a shrouded impeller, which is partly why the developers of the Levacor chose a semi-open design (55). The high shear regions seem to be related to leakage of turbulent flow from the interblade space, because some of the highest shear regions in figure 19A correspond to regions of increased turbulence in figure 13A.

4.1 Implications for Endothelial Cell Attachment to LVAD Walls:
Our collaborator at Duke Medical Center can currently adhere a confluent lining of endothelial progenitor cells to titanium under 10Pa of shear stress and has achieved 90% endothelial progenitor cell retention at shear stresses as high as 20-30Pa (unpublished data). However, the shear stresses in figures 15 through 19 are still too high for endothelialization to be realistic on most walls of this particular centrifugal LVAD. Only the side wall of the upper volute experiences less than 10Pa of shear stress over more than 10% of its surface (see Figure 16). The low-shear area on this wall is mainly near the outlet, on the side furthest from the impeller. Because shear stress is above 50Pa on the bottom of the lower volute and above 100Pa on its side walls, endothelialization of this region is particularly unlikely. Furthermore, endothelial lining the lower volute would make the already narrow flow path between the impeller and lower walls even narrower. In an early version of the HeartMate II, growth of pseudointima in narrow regions near the inlet and outlet stators was cited as a possible cause of
primary pump thrombosis (56). Replacing the sintered surface that encouraged growth of pseudoneointima in these areas with smooth surfaces reduced the incidence of thrombus. However, high pressure in the narrow secondary flow paths of the HeartMate III may prevent thrombus from occurring in this case. The upper volute generally has lower shear stress, and the region nearest the outlet has wall shear stresses lower than 20Pa for the both LVAD designs. This region the best possible site for endothelialization in the LVADs studied here. Additionally, the inlet pipe of the HeartMate III, which was not simulated here, may be a good candidate for endothelialization. Burgreen and colleagues found low flow velocity in the inlet elbow region (52). Overall, this LVAD is not ideal for endothelialization. A pump designed specifically with endothelial lining in mind, with wider flow paths and lower shear stresses, may be preferable. The HeartMate III, along with many other LVADs, is designed so that high shear flows will wash the inner walls, preventing thrombus. These high shear flows also make endothelialization difficult, but an endothelialized LVAD would not need high shear stresses if the endothelial lining is successful in preventing thrombus.

4.2 Implications for Von Willebrand Factor:

The radial gap between the impeller and wall of the volute has shear rates well over 10,000/s, which corresponds to 35Pa (350dyn) of shear stress. This value is easily high enough to extend long VWF multimers and expose cleavage sites. However, it is currently unclear if the average residence time of less than 5ms in
this region is sufficient to allow for VWF cleavage, because the time needed for ADAMTS13 proteases to bind VWF multimers and cleave them is not known. One region of concern is the lower part of the central core within the impeller’s lower shroud. This section has slightly higher shear rates than other parts of the central core (Figure 22). More blood is likely exposed to this region for a longer time period than the radial gap between the impeller and lower volute wall, which could leave VWF vulnerable to cleavage.

Figures 19 and 21 indicate slightly higher shear above the semi-open impeller away from the outlet side of the LVAD. Figure 13 also indicates that this region has higher turbulent kinetic energy in the LVAD with a semi-open impeller. It appears that, the semi-open impeller allows turbulent flow to escape from the inter-blade region, which may have contributed to the higher shear stress there. In the other LVAD design, the shroud would have blocked this turbulent flow from escaping the inter-blade region vertically. Further investigation of this region and the central core region is necessary to determine whether the shrouded or semi-open design is preferable for preventing degradation of large VWF multimers. Additionally, shear rate and residence time in high-shear regions should be investigated at lower, more clinically typical flow rates than the 7L/min flow rate used here. Lower flow rates may decrease shear and increase residence time, which could change how VWF multimers are affected.
4.3 Limitations and Future Directions:

In this study, we have analyzed shear rate and residence time in high shear regions of a centrifugal LVAD with a particular focus on how these may affect VWF. However, although research on VWF is progressing rapidly, the precise threshold combination of residence time and shear rate that results in VWF degradation is currently unknown. As the molecular dynamics of VWF multimers in shear flow become clearer through further research, the CFD results presented here will become more useful.

This model has been somewhat limited by computational expense and technical constraints. Combined with the $k - \omega$ model and sliding mesh model, the mesh used in this study is so computationally expensive that it must be simulated on a supercomputer. The number of impeller rotations that could be simulated was limited by supercomputer time. While a fifth rotation of the shrouded impeller produced results that were similar to those after the fourth rotation, it was too computationally expensive to continue simulating rotations until flow data converged to an entirely consistent periodic flow from one impeller rotation to the next. Therefore, we cannot say with certainty that the results presented here represent the exact behavior of the LVAD in the body after thousands of impeller rotations.

In a previous study of a similar centrifugal LVAD, the results from CFD analysis largely agreed with experimental results, but minor variations of less than 10%
existed (52). In this study, we have no experimental results for comparison, but we suspect that these results differ slightly from experimental data as well. The main contributor to this discrepancy is numerical diffusion. Resulting from truncation errors that are caused by representing continuous flow equations in discrete form, numerical diffusion is present with all numerical solving schemes. Although we have attempted to reduce numerical diffusion by using second order discretization and hexahedral mesh volumes in areas of simple flow such as the outflow tube, this source of error cannot be entirely eliminated.

Our future research will analyze different flow rates through this LVAD. The current flow rate of 7L/min was chosen here to allow easier comparison to the CFD analysis of the HeartMate III by Burgreen and colleagues, who studied a 7L/min flow (52). However, 7L/min is a higher blood flow rate than is normally provided by a healthy heart and is also higher than what most LVADs produce clinically (15), (56). Therefore, it will be useful to attempt a lower flow rate that is more clinically realistic such as 4 or 5L/min. These lower flow rates may result in lower shear stresses and a more hospitable environment for adhesion of endothelial progenitor cells to the LVAD walls.

4.4 Conclusions:

The LVAD studied here is largely inappropriate for endothelialization regardless of which impeller type is used. The only areas with shear stresses low enough to allow endothelialization are the walls closest to the outlet, and a centrifugal pump
designed specifically to create lower shear stresses would better support endothelial lining. The highest shear rates occur in the gap between the impeller and lower volute wall, but residence times below 5ms make cleavage of large VWF multimers less likely in this region. Other high-shear areas exist in the central core, and these regions require further study to better understand the risk of VWF cleavage.
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Appendix A: Turbulence

The formulas and information presented in this appendix are largely based on chapter four of the ANSYS FLUENT Theory Guide. The k-ε model is the most commonly used turbulence model and has been used in several CFD studies of LVADs (45), (55). However, because this model assumes that flow is fully turbulent, it works best at very high Reynolds numbers and is only appropriate for fully turbulent flows. One option for dealing with lower Reynolds numbers is The RNG k-ε Model.

A.1 The RNG k-ε Model

The RNG k-ε model applies a statistical technique called renormalization group theory to the instantaneous Navier-Stokes equations, resulting in equations that are slightly different than in the k-ε model.

\[
\begin{align*}
\frac{\partial}{\partial t} (\rho k) + \frac{\partial}{\partial x_i} (\rho ku_i) &= \frac{\partial}{\partial x_j} \left( \alpha_k \mu_{\text{eff}} \frac{\partial k}{\partial x_j} \right) + G_k + G_b - \rho \varepsilon - Y_M \quad (A - 1) \\
\frac{\partial}{\partial t} (\rho \varepsilon) + \frac{\partial}{\partial x_i} (\rho \varepsilon u_i) &= \frac{\partial}{\partial x_j} \left( \alpha_{\varepsilon} \mu_{\text{eff}} \frac{\partial \varepsilon}{\partial x_j} \right) + C_{1\varepsilon} \frac{\varepsilon}{k} (G_k + C_3 \varepsilon G_b) - C_{2\varepsilon} \rho \frac{\varepsilon^2}{k} - R_\varepsilon \\
\end{align*}
\]

where the RNG-derived constants \( C_{1\varepsilon} = 1.42 \), \( C_{2\varepsilon} = 1.68 \), and the inverse effective Prandtl numbers \( \alpha_k \) and \( \alpha_\varepsilon \) are calculated with this formula, also derived from RNG theory:

\[
\begin{align*}
\frac{\alpha - 1.3929}{\alpha_0 - 1.3929}^{0.6321} \quad \frac{\alpha + 2.3929}{\alpha_0 + 2.3929}^{0.3679} &= \frac{\mu_{\text{mol}}}{\mu_{\text{eff}}} \quad (A - 3)
\end{align*}
\]
where $\alpha_0 = 1.0$ and $\mu_{\text{mol}} =$ molecular viscosity. For high Reynolds numbers, $\alpha_k = \alpha_\varepsilon \approx 1.393$. The remaining term that distinguishes the RNG model from the standard model is $R\varepsilon$.

\[
R\varepsilon = \frac{C_\mu \rho \eta^3 (1 - \eta / \eta_0) \varepsilon^2}{1 + \beta \eta^3} k \tag{A - 4}
\]

where

\[\eta \equiv S k / \varepsilon, \quad \eta_0 = 4.38, \quad \beta = 0.012\]

In regions of high strain rate, where a large $S$ term yields $\eta > \eta_0$, the $R$ term is reduced, meaning there is less destruction of turbulent dissipation $\varepsilon$ in equation A - 2, giving a higher $\varepsilon$ and lower $k$. This results in lower turbulent viscosity in highly strained flows and more accurate responsiveness to rapid strains in the RNG model.

In the RNG $k$-$\varepsilon$ model, turbulent viscosity is solved either by

\[
\mu_t = \rho C_\mu \frac{k^2}{\varepsilon} \tag{A - 5}
\]

or by integrating the equation

\[
d \left( \frac{\rho^2 k}{\sqrt{\varepsilon \mu}} \right) = 1.72 \frac{\hat{\nu}}{\sqrt{\hat{\nu}^3 - 1 + C_{\nu}}} d\hat{\nu} \tag{A - 6}
\]

where $\hat{\nu} = \mu_{\text{eff}} / \mu$ and $C_{\nu} \approx 100$.

Equation A - 6 describes how turbulent transport varies with Reynolds numbers.

At high Reynolds numbers, the solution to equation A - 6 approaches equation A - 5 with $C_\mu = 0.0845$. In Fluent, A - 5 is the default turbulent viscosity equation.
for the RNG k- ε model. However, selecting the Differential Viscosity Model option enables the use of equation A - 6, allowing for more accurate determination of turbulent viscosity in areas of low-Reynolds-number flows.

**A.2 The k – ω Model**

The standard k - ω model has modifications for low-Reynolds-number turbulence and can therefore be used accurately for free shear and wall-bounded flows. It is defined by the following transport equations:

\[
\frac{\partial}{\partial t} (\rho k) + \frac{\partial}{\partial x_i} (\rho k u_i) = \frac{\partial}{\partial x_j} \left( \Gamma_k \frac{\partial k}{\partial x_j} \right) + G_k - Y_k \tag{A - 7}
\]

\[
\frac{\partial}{\partial t} (\rho \omega) + \frac{\partial}{\partial x_i} (\rho \omega u_i) = \frac{\partial}{\partial x_j} \left( \Gamma_\omega \frac{\partial \omega}{\partial x_j} \right) + G_\omega - Y_\omega \tag{A - 8}
\]

\(G_k\) and \(G_\omega\) are generation of \(k\) and \(\omega\), respectively, and \(Y_k\) and \(Y_\omega\) are dissipation of turbulence and \(\omega\), respectively. The generation terms and effective diffusivities \(\Gamma_k\) and \(\Gamma_\omega\) can be found with the following equations

\[
\Gamma_k = \mu + \frac{\mu_t}{\sigma_k} \tag{A - 9}
\]

\[
\Gamma_\omega = \mu + \frac{\mu_t}{\sigma_\omega} \tag{A - 10}
\]

\[
G_k = -\rho u_i u_j \frac{\partial u_j}{\partial x_i} \tag{A - 11}
\]

\[
G_k = \mu_t s^2 \tag{A - 12}
\]
\[ G_\omega = \alpha \frac{\omega}{k} G_k \]  
(A - 13)

\[ \alpha = \frac{\alpha_\infty}{\alpha^*} \left( \frac{\alpha_0 + R e_t / R_\omega}{1 + R e_t / R_\omega} \right) \]  
(A - 14)

where

\[ R_\omega = 2.95. \]

The equations for the dissipation terms \( Y_k \) and \( Y_\omega \) can be found in the Fluent Theory Guide, Section 4.5.

The low-Reynolds-number correction is provided by the coefficient \( \alpha^* \) in the following equation for \( \mu_t \).

\[ \mu_t = \alpha^* \frac{\rho k}{\omega} \]  
(A - 15)

\[ \alpha^* = \alpha^*_\infty \left( \frac{\alpha_0^* + R e_t / R_k}{1 + R e_t / R_k} \right) \]  
(A - 16)

where

\[ R e_t = \frac{\rho k}{\mu \omega} \]  
(A - 17)

\[ R_k = 6 \]  
(A - 18)

\[ \alpha_0^* = \frac{\beta_i}{3} \]  
(A - 19)

\[ \beta_i = 0.072 \]  
(A - 20)

The constants used in section A.2 are found below.

\[ \alpha_\infty = 1, \ \alpha_\infty = 0.52, \ \alpha_0 = \frac{1}{9}, \ \beta^*_\infty = 0.09, \ \beta_i = 0.072, \ R_\beta = 8 \]

\[ R_k = 6, \ R_\omega = 2.95, \ \zeta^* = 1.5, \ M_{i0} = 0.25, \ \sigma_k = 2.0, \ \sigma_\omega = 2.0 \]
A.3 The SST $k - \omega$ Model

As discussed previously, the $k - \omega$ model is more accurate for near-wall flows and other situations with incomplete turbulence, while the $k - \varepsilon$ model is more appropriate for fully turbulent free stream flows. The advantage of the SST $k - \omega$ model contains elements of both the $k - \omega$ and $k - \varepsilon$ models. The benefit of the SST model is that it creates a gradual change from the $k - \omega$ model very close to a wall to the $k - \varepsilon$ model in the outer boundary layer and free stream. This gradual transition is facilitated by the blending functions $F_1$ and $F_2$ found in Equations A – 28 through A – 31 below. Melding the two models allows the SST model to be more accurate for a wider range of flows including shockwaves and airfoils.

The equations to solve for $k$, $\omega$ and effective diffusivities for this SST model have forms similar to their counterparts in the standard $k - \omega$ model.

\[
\frac{\partial}{\partial t} (\rho k) + \frac{\partial}{\partial x_i} (\rho k u_i) = \frac{\partial}{\partial x_j} \left( \Gamma_k \frac{\partial k}{\partial x_j} \right) + \tilde{G}_k - Y_k + S_k \quad (A - 21)
\]

\[
\frac{\partial}{\partial t} (\rho \omega) + \frac{\partial}{\partial x_i} (\rho \omega u_i) = \frac{\partial}{\partial x_j} \left( \Gamma_\omega \frac{\partial \omega}{\partial x_j} \right) + G_\omega - Y_\omega + D_\omega + S_\omega \quad (A - 22)
\]

\[
\Gamma_k = \mu + \frac{\mu_t}{\sigma_k} \quad (A - 23)
\]

\[
\Gamma_\omega = \mu + \frac{\mu_t}{\sigma_\omega} \quad (A - 24)
\]
However, terms such as turbulent viscosity and Prandtl numbers (\(\sigma_k\) and \(\sigma_\omega\)) in the SST equations above are calculated differently than in the standard model.

\[
\mu_t = \frac{\rho k}{\omega \max \left[ \frac{1}{\alpha^+}, \frac{S_F}{a_1 \omega} \right]}
\]

\[
\sigma_k = \frac{1}{F_1/\sigma_{k,1} + (1 - F_1)/\sigma_{k,2}}
\]

\[
\sigma_\omega = \frac{1}{F_1/\sigma_{\omega,1} + (1 - F_1)/\sigma_{\omega,2}}
\]

where

\[
F_1 = \tanh \left( \Phi_1^4 \right)
\]

\[
\Phi_1 = \min \left[ \max \left( \frac{\sqrt{k}}{0.09\omega y}, \frac{500\mu}{\rho y^2 \omega} \right), \frac{4\rho k}{\sigma_{\omega,2}D_\omega^+ y^2} \right]
\]

\[
D_\omega^+ = \max \left[ 2\rho, \frac{1}{\sigma_{\omega,2}} \omega \frac{\delta k}{\partial x_j} \frac{\delta \omega}{\partial x_j}, 10^{-10} \right]
\]

\[
F_2 = \tanh \left( \Phi_2^2 \right)
\]

\[
\Phi_2 = \max \left[ \frac{2\sqrt{k}}{0.09\omega y}, \frac{500\mu}{\rho y^2 \omega} \right]
\]
In equation A – 25, S is the magnitude of strain rate, and, in equations A – 29 and A - 32, y is the distance to the nearest surface.

\[ \tilde{G}_k = \min(G_k, 10\rho\beta^* k\omega) \]  
(A - 33)

\[ G_\omega = \frac{\alpha}{\nu_t} \tilde{G}_k \]  
(A - 34)

The turbulent viscosity (equation A – 25) is modified from its standard k - ω model form (equation A – 15) to account for transport of turbulent shear stress. Note that, while \( \alpha \) is still defined by equation A – 14, \( \alpha_\infty \) is here defined using one of the blending functions.

\[ \alpha_\infty = F_1\alpha_{\infty,1} + (1 - F_1)\alpha_{\infty,2} \]  
(A - 35)

where

\[ \alpha_{\infty,1} = \beta_{i,1} - \frac{\kappa^2}{\beta_\infty^*} \frac{\sigma_{w,1}}{\sigma_{w,1}} \sqrt{\beta_\infty^*} \]  
(A - 36)

\[ \alpha_{\infty,2} = \beta_{i,2} - \frac{\kappa^2}{\beta_\infty^*} \frac{\sigma_{w,2}}{\sigma_{w,2}}\sqrt{\beta_\infty^*} \]  
(A - 37)

The equations for the dissipation terms \( Y_k \) and \( Y_\omega \) are the same here as in the standard model, but

\( f_{\beta^*} = f_\beta = 1 \) and \( \beta_i \) is no longer constant.

\[ \beta_i = F_1\beta_{i,1} + (1 - F_1)\beta_{i,2} \]  
(A - 38)
\[ F_1 = \tanh \left( \Phi_1^4 \right) \]  

Because the SST model combines the \( k - \varepsilon \) and \( k - \omega \) models, the \( k - \varepsilon \) model must be converted into terms of \( k \) and \( \omega \). This introduces a cross-diffusion term \( D_\omega \) in Equation A – 22.

\[
D_\omega = 2 \left( 1 - F_1 \right) \rho \sigma_{\omega,2} \frac{1}{\omega} \frac{\partial k}{\partial x_j} \frac{\partial \omega}{\partial x_j} \quad (A - 40)
\]

The constants used in A – 3 that differ from those used in A – 2 are found below.

\[
\sigma_{k,1} = 1.176, \quad \sigma_{\omega,1} = 2.0, \quad \sigma_{k,2} = 1.0, \quad \sigma_{\omega,2} = 1.168 \\
a_1 = 0.31, \quad \beta_{i,1} = 0.075 \quad \beta_{i,2} = 0.0828
\]
Appendix B: Data after Three Impeller Rotations

The following plots are on data gathered after three complete impeller rotations and are given for comparison to the data presented in the results section.

Figure B1: Fluid Velocity Magnitude (m/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure B2: Axial Velocity (m/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure B3: Radial Velocity (m/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure B4: Radial Velocity on a Transverse Mid-plane Cut through the Vertical Center of the Impeller Blades (m/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure B5: Static Gauge Pressure (Pa) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller. Inlet Pressure Set at Zero.
Figure B6: Tangential Velocity on a Transverse Mid-plane Cut through the Center of the Impeller Blades (m/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure B7: Tangential Velocity on a Transverse Mid-plane Halfway between the Impeller and Top of the Volute (m/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure B8: Turbulent Kinetic Energy ($m^2/s^2$) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure B9: Turbulent Kinetic Energy on a Transverse Mid-plane Halfway between the Impeller and the Top of the Volute (m²/s²). A) Semi-open Impeller; B) Shrouded Impeller
Figure B10: Velocity Vectors (m/s) Colored by Axial Velocity with Eddies Circled on Vertical Plane through the Center of the LVAD. A) Semi-open Impeller; B) Shrouded Impeller
Figure B11: Shear Stress on Side Wall of Lower Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure B12: Shear Stress on Side Wall of Upper Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure B13: Shear Stress on Bottom Wall of Lower Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure B14: Shear Stress on Bottom Wall of Upper Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure B15: Shear Stress on Top Wall of Volute (Pa). A) Semi-open Impeller; B) Shrouded Impeller
Figure B16: Shear Rate on a Transverse Mid-plane Cut through the Center of the Impeller Blades (1/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure B17: Shear Rate on a Transverse Plane Midway between the Impeller and the Top of the Volute (1/s). A) Semi-open Impeller; B) Shrouded Impeller
Figure B18: Shear Rate (1/s) on Vertical Plane Cut through the Center of LVAD with A) Semi-open Impeller; B) Shrouded Impeller
Figure B19: Magnified View of Shear Rates in Radial Blade Clearance Gaps (1/s).
A) Semi-open Impeller; B) Shrouded Impeller