Effect of LVAD Outlet Graft Anastomosis Angle on the Aortic Valve, Wall, and Flow

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Left ventricular assist devices (LVADs), which pump blood from the left ventricle to the aorta are an important therapy option for patients with end-stage cardiovascular diseases. Recent publications show that even with optimized LVADs fatal complications can occur because of the blood deformations around the inflow cannula or through the LVAD outlet graft–aorta anastomosis. This study investigates the effects of the anastomosis geometry on the flow through the aorta, on the pressure and wall shear stress (WSS) distributions on the aortic wall and on the total entropy generation in the anastomosis region. Anastomosis geometry is defined with two angles, one on the coronal plane and the other on the transversal plane. Turbulent flow simulations are performed for each geometry. Results indicate that 3% to 5% of the work given by the LVAD is dissipated because of the viscous losses in the anastomosis region. The entropy generation, as well as the maximum WSS, increases as the inclination angle decreases. Some portion of the blood streaming out of the LVAD conduit flows toward the aortic valve; therefore the reverse-flow region extends up to the aortic valve in some cases, which may be one of the causes of aortic-valve dysfunction. Results of this study provide insight on the importance of the anastomosis geometry on the hemodynamics in the aorta and downstream the aortic valve, stresses on the aortic wall, and viscous losses. ASAIO Journal 2012;58:373–381.

Cardiovascular diseases and resulting deaths are the most frequent cause in global mortality with approximately 17.1 million deaths worldwide, i.e., 29% of all deaths in 2004. Congestive heart failure, which makes up a considerably large portion of cardiovascular diseases, is a condition in which the heart fails to provide enough blood to the body. A promising option is the left ventricular assist device (LVAD). The inflow conduit of the LVAD, which is connected to the apex of left ventricle, pulls blood from the left ventricle into a pump, whereas the outflow conduit is connected to the aorta to enable the oxygen-rich blood circulation throughout the body (Figure 1).

A myriad of studies have been done to increase the performance of LVADs as well as to anticipate complications that could occur. Many studies, however, only pinpoint to the blood flow inside the heart pumps. Because of the high rotational speeds of pumps, blood cells may pass through the pump without excessive deformation or hemolytic damage. Clinical investigations show that the inlet and outlet conduits of an LVAD severely affect the hemodynamics in the left ventricle and the native aorta, respectively. This may lead to significant complications such as thrombosis, platelet activation, and aneurysm in the aorta. In addition, clinical experience shows that the aortic valve may be damaged as a consequence of an LVAD implantation, because of the altered hemodynamics upstream and downstream the aortic valve. Computational fluid dynamics (CFD) could be a useful tool in investigating the altered hemodynamics and their effects on the aortic valve and wall.

In patients with an LVAD, flow rate through the LVAD ranges between 2.5 and 7.5 L/min and through the aortic valve between 0 and 5 L/min. If we assume average values for flow rates, Reynolds number at the aorta inlet is calculated as 520 and at the LVAD conduit outlet as 2430. Blood flow through the LVAD–aorta anastomosis contains laminar regimes as well as transitional regimes. To accurately predict the variables in this flow, the numerical method should be able to simulate laminar, transitional, and turbulent regimes simultaneously. An interlaboratory study performed by Food and Drug Administration showed the importance of turbulence modeling in low Reynolds number flows. Even for a nozzle flow, where the maximum Re was 500, the standard deviation (SD) of the CFD data from 28 different groups was greater than 60% of the mean value. At low Reynolds numbers k-ω and k-ω shear stress transport (SST) models provide more accurate results than the k-ε model, which tends to overpredict turbulent kinetic energy.

Only a few studies are published on the aorta–LVAD outlet graft anastomosis. May-Newman et al. investigated effects of the location and the insertion angle of the LVAD outlet graft by idealizing the aorta geometry as a straight tube or as a three-dimensional arc assuming laminar flow. Kar et al. performed an unsteady simulation of the anastomosis by modeling aorta as a two-dimensional arc. Aortic blood flow caused by implantation of the Jarvik 2000 LVAD (Jarvik Heart, Inc., New York, NY) is investigated by Bazilevs et al. with a patient-specific aorta, where the outlet graft is connected to the descending aorta. They performed the simulations with a residue-based variational multiscale turbulence model, and observed turbulent structures in the aorta.

This article investigates the altered hemodynamics in a patient-specific aorta, when the LVAD outlet is connected to the ascending aorta via CFD. When the anastomosis is positioned...
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in the ascending aorta, the jet flow from the outlet graft affects both the pressure on the aortic valve and flow rate through the subclavian arteries greatly. Simulations are performed with k-ω SST model to predict the laminar–turbulent transitional accurately. Anastomosis is described by the connection angles on the coronal and transversal planes. The effect of the anastomosis geometry on the flow through the aorta, on the pressure and wall shear stress (WSS) on the aortic wall and the aortic valve are investigated. Entropy generation caused by the viscous losses in the anastomosis region is estimated.

Methods

Geometry

Three-dimensional aorta geometry was extracted from the computer tomography (CT) scan of a 65 year old male volunteer who previously had undergone a bypass surgery. The scan contains approximately 500 slices with a distance of 0.625 mm. The boundary of the aortic wall is reconstructed from these slices via image processing (Figure 2). Anatomic structures other than the aorta, such as the heart, small vessels, and bones are filtered out to obtain an image that contains only the aorta geometry. The arteries that branch out of the aortic arch, i.e., subclavian, brachiocephalic, and carotid arteries, are left out of the flow domain. A mesh of triangular cells is fitted on the inner surface of the aorta, and a network of curves is sketched based on this triangular surface mesh. These curves are used to define several surfaces that enclose the flow domain. The flow domain contains the aortic arc, ascending, and descending aorta.

The common surgical procedure is to connect the LVAD outlet graft to the ascending aorta. Because the location of a previous bypass surgery on the ascending aorta is visible in the CT scans, this point is chosen as the anastomosis location.

The LVAD outlet graft is represented as a rigid tube with a diameter of 12 mm. The direction of the outlet graft is described in spherical coordinate system with the azimuthal (θ) and inclination (ϕ) angles (Figure 3). Anastomosis with three different azimuthal angles (90°, 100°, and 120°) and six different inclination angles (15°, 30°, 45°, 60°, 75°, and 90°) were studied.

Flow Simulation

Blood is modeled as an incompressible (ρ = 1,050 kg/m³) and Newtonian fluid (μ = 0.0035 Pa s). Aorta and LVAD graft are modeled as rigid walls. Parallel flow through the aorta and LVAD is assumed, so that 2.5 L/min and 5 L/min are set at the aorta and graft inlet, respectively. Uniform velocity perpendicular to the inlet surface is set at the inlet boundaries. At the outlet static pressure is set as 0 gage pressure and all the other variables are extrapolated from the fluid domain. Reynolds Averaged Navier-Stokes (RANS) simulations are performed for all designs with the software Fluent 6.2 (Ansys Inc., Canonsburg, PA). The continuity and the RANS equations are given below.

\[
\frac{\partial u_i}{\partial x_i} = 0 \tag{1}
\]

\[
\frac{\partial (\rho u_i u_j)}{\partial x_j} = \frac{\partial p}{\partial x_j} - \frac{\partial \tau_{ij}}{\partial x_j} - \frac{\partial (\rho u_i u_j)}{\partial x_j} + \rho g_i \tag{2}
\]

The closure problem, which arises because of the emergence of the Reynolds stress tensor, is solved with the Boussinesq approximation by defining the eddy viscosity \( \mu_t \).
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Reynolds number in the LVAD outlet graft is $Re_{\text{graft}} = 2,430$ and in the aorta inlet is $Re_{\text{valve}} = 520$. Although the $Re$ is low, turbulent structures may occur locally especially in the anastomosis area because of the sudden geometry change and mixing of two flow streams. To simulate laminar, transitional, and turbulent regions adequately, the $k$-$\omega$ SST model is employed. The transport of $k$ and $\omega$ are calculated with the following partial differential equations:

$$\rho \frac{\partial u_i}{\partial x_j} = -\rho \left( \frac{\partial \sigma}{\partial x_i} + \frac{\partial \sigma}{\partial x_j} \right) + \frac{2}{3} \delta_{ij} \rho k$$  \hspace{2cm} (3)

$$\frac{\partial (\rho k)}{\partial x_j} = \frac{\partial}{\partial x_j} \left[ \left( \frac{\mu_1}{\sigma_k} \frac{\partial k}{\partial x_j} \right) + C_k - Y_k + \Pi_k \right]$$  \hspace{2cm} (4)

Where, $C_k$ is the generation of $k$ because of the mean velocity gradients ($C_k = \mu_1 \frac{2}{3} \mathbf{S} \mathbf{S} \mathbf{S}$, where $\mathbf{S}$ is the strain tensor). $S_x$ and $S_y$ are zero here. The governing equations are discretized with second-order upwind model and solved with segregated-pressure based implicit method.

**Grid Independency**

To assure grid independent data, numerical results of three structured O-type grids are compared. The total number of control volumes in these meshes is approximately 520,000, 2 million, and 4.5 million, respectively (Figure 4A). To provide a fair comparison, all the model settings (turbulence model,}

![Figure 3. Azimuthal and inclination angles on the transversal and coronal planes defining the anastomosis direction.](image)

![Figure 4. A: Computational meshes for grid dependency test, (B) velocity vectors calculated with different computational meshes.](image)
boundary conditions, convergence criteria, and discretization scheme etc.) are kept the same.

Numerical results were evaluated considering the total pressure drop, WSS distribution on the aortic wall, pressure distribution on the middle plane, and secondary flows in the descending aorta (Figure 4B). The difference between the finest and moderate meshes in the total pressure drop decreased below 5% and in the average WSS decreased below 2%. The finest mesh was used for the rest of the simulations.

**Results and Discussion**

If the LVAD graft is modeled as a straight tube, then the area of the anastomosis increases with decreasing $\phi$, and becomes up to four times larger than the graft cross-sectional area (Figure 5). Furthermore, velocity vectors at the graft outlet become parallel to $\phi$. Correspondingly, high-velocity magnitudes appear in the aortic arch, which may lead to physiologically unrealistic flow rates through the subclavian arteries. These problems are avoided by twisting the distal end of the graft. As a result, changes in $\phi$ effects primarily the hemodynamics within the graft. Another result of twisting the distal end is that the inclination angle of the velocity vectors at the graft outlet decreases and in the ascending aorta reverse flow toward the aortic valve is formed. For both geometries, entrance and outflow regions are laminar, but turbulence is generated because of the sudden geometry change and mixing of flows from the left ventricle and LVAD (Figure 5, C and D).

Flow from the LVAD outlet graft resembles a jet flow and generates vorticity as it hits the aortic wall (Figure 6). Bulk flow from the LVAD outlet is directed toward the aortic arc. Vortices formed between the anastomosis and the aortic arc dissipate toward the end of the arc, for low $\phi$. As $\phi$ grows to 90°, vorticity and helicity remain dominant and effect even the flow in the descending aorta.

Although bulk flow is toward the arc, a portion of the blood from the LVAD outlet flows toward the aortic valve. Large vortices between the anastomosis and aortic valve are visible, especially at low $\theta$ and high $\phi$. The secondary flow structures, which are generated by the LVAD left ventricular flow interaction and located proximal to the ascending aorta may possibly blockade flow from the aortic valve (e.g., configuration $\theta = 90^\circ$ and $\phi = 90^\circ$ in Figures 6 and 8).

At $\theta = 90^\circ$, the jet flow from the LVAD outlet directly flows through the center of the aorta. The jet stream hits the aortic wall and Dean-like vortices appear (Figure 7). At larger azimuthal angles vortices in the secondary flow vanish. At 120° the streamlines from the LVAD outlet follow the curvature of the aortic wall and are directed to the aortic arc. This suppresses the Dean-like vortices.

**Figure 5.** Contours of the velocity magnitude on a vertical cross-section (A) if the LVAD graft is modeled as a straight tube, (B) if the LVAD graft is twisted at the end. Contours of the turbulent kinetic energy (C) if the LVAD graft is modeled as a straight tube, (D) if the LVAD graft is twisted at the end. (Here $\theta = 90^\circ$ and $\phi = 30^\circ$). LVAD, left ventricular assist device.
The importance of the entrance region on the local hemodynamics is demonstrated further with Figure 8. Cross-sectional plane on which velocity vectors are shown rotates with the azimuthal angle to show the complex rotational structures better. Large-velocity gradients near the aortic wall and reverse-flow regions upstream and downstream the anastomosis are visible in each configuration. Reverse flow even toward the LVAD graft is visible at low $\theta$. The influence of graft flow (i.e., reverse flow between the anastomosis and aortic valve) is minimum for high $\theta$.

High WSS values appear at the toe of the anastomosis (Figure 9). In each configuration, a local minimum appears between two hot spots. Average WSS ($WSS_{\text{ave}}$) values on the conduit wall are higher than $WSS_{\text{ave}}$ on the aortic wall, because the conduit diameter is smaller and the mass flow rate through the conduit is higher. Surface averaged WSS values exhibit small sensitivity to anastomosis angles. $WSS_{\text{ave}}$ ranges between 1.1 and 1.2 Pa on the aortic wall, whereas it ranges between 5.0 and 7.2 Pa on the conduit wall. The maximum WSS ($WSS_{\text{max}}$) on the conduit wall too is not correlated to the...
anastomosis angles. It remains nearly constant at 71.4 ± 0.4 Pa for each configuration.

Generally, $WSS_{\text{max}}$ decreases as $\theta$ decreases, or as $\phi$ increases (Figure 10A), although there are some exceptions, e.g., a small rise in $WSS_{\text{max}}$ occurs as $\phi$ rises from 75° to 90°. Results show that the effect of $\phi$ is larger than that of $\theta$.

In general, as $\phi$ increases, the maximum pressure increases (Figure 10B). However, at $\theta = 90^\circ$, the increase with respect to $\phi$ is negligibly small (only 40 Pa). At $\theta = 120^\circ$, the maximum pressure remains nearly constant for low inclination angles. As $\phi$ increases beyond 45°, the maximum pressure increases steeply. Maximum pressure is 220 Pa higher at 90° than at 45°.

From a thermodynamic point of view the entire cardiovascular system (as shown in Figure 1) can be regarded as a closed system. LVAD takes electrical power from a motor and increases the energy of the blood by increasing the blood pressure. This energy is consumed to overcome the viscous losses. The second law of thermodynamics allows us to estimate the irreversibilities in a process. The second law of thermodynamics for a closed, adiabatic and steady system can be written as\(^{17}\):

$$\eta W_{\text{LVAD}} = \tau \dot{\gamma}_{\text{m}}$$

(6)

Figure 8. Three-dimensional velocity vectors on vertical cross-sections.

Figure 9. WSS distributions on the aortic and conduit wall. WSS, wall shear stress.
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Here, \( T_0 \) is 36.5°C. If we assume that the employed LVAD provides a pressure increase of \( \Delta p_{LVAD} = 100 \) mmHg at a flow rate of \( V = 5 \) L/min (which is usually chosen as the operating point during the design of LVADs), then the first term of the equation 6 becomes:

\[
\eta V_{pump} = \dot{V} \Delta p_{LVAD} = 1.1 \text{ J/s} \tag{7}
\]

This available energy is consumed to overcome the viscous losses inside the cardiovascular system. The dominant source of entropy generation is the immense total length of the blood vessels. A small part of the available energy is consumed as a result of the losses in the anastomosis region. The entropy generated in the anastomosis region can be calculated as:

\[
\dot{S}_{gen, anastomosis} = (\rho \dot{V} s)_{aorta outlet} - (\rho \dot{V} s)_{aorta inlet} - (\rho \dot{V} s)_{conduit inlet} \tag{8}
\]

The subscripts refer to the inlet and outlet boundaries chosen for the flow simulation (Figure 1). Entropy at a thermodynamic state can be calculated as the sum of the entropy at a reference state and the change of entropy between the thermodynamic and reference states:

\[
s = s_0 + \Delta s \tag{9}
\]

The reference state can be defined arbitrarily. Let us choose the outflow boundary conditions of the CFD flow domain as our reference state, i.e., \( s_0 = s_{aorta outlet} \) and \( \rho_0 = \rho_{aorta-outlet} \). The entropy change for an incompressible fluid is:

\[
ds = -\frac{1}{\rho_0} \int \frac{dp}{\Gamma} \tag{10}
\]

Accordingly:

\[
\Delta s = \int_{s_0}^{s} ds = -\int_{s_0}^{s} \frac{1}{\rho_0} \frac{dp}{\Gamma} \tag{11}
\]

If the integration for the conduit and aorta inlets is carried out, and the results are inserted into equation 8, then \( \dot{S}_{gen, anastomosis} \) becomes:

\[
\dot{S}_{gen, anastomosis} = \frac{(\dot{V} p)_{aorta inlet} + (\dot{V} p)_{conduit inlet}}{T_0} \tag{12}
\]
An interesting result of this analysis is that the entropy generation does not change with \( \theta \) (Figure 10C). Although the flow structure (vorticity, WSS distributions, and streamlines etc.) are affected by the changes in \( \theta \), the overall viscous losses do not depend on \( \theta \). A monotone decrease in the entropy generation is observed as \( \phi \) increases. Depending on \( \phi \), 3\% to 5\% of the total entropy loss occurs through the anastomosis region.

**Conclusion**

LVAD outlet graft and aorta anastomosis is modeled with the azimuthal and inclination angles. Eighteen flow simulations were performed to compare the effect of the anastomosis angle on the altered hemodynamics in the aorta, pressure, and WSS distributions on the walls of the aorta and the conduit and the irreversibilities that occurred because of the viscous losses.

The main limitation of this study is the lack of validation. Visualization of the real flow with techniques like magnetic resonance imaging is hardly possible. Our further research focuses on experimental measurements of this flow either via flow visualization with smoke by scaling the geometry up approximately 10 to 15 times, or via particle image velocimetry (PIV). PIV may provide a basis for validation, although PIV of physiological flows are bound to have important experimental errors as exemplified in the study by Hariharan et al.\(^{10} \) Omitting the arteries that branch out of the aortic arch, i.e., subclavian, brachiocephalic, and carotid arteries, makes the current model incapable of estimating the effect of graft orientation on the aortic arch and descending aorta flow structures. Another limitation of the study is the rigid wall assumption. Shape and length of the aorta, LVAD outlet graft, and anastomosis region would change because of time-dependent wall pressure, shear forces, and muscle reactions. Correspondingly, hemodynamics would change too. However, this effect is neglected, because it lies beyond the scope of this study. The accuracy of the results shown in this article may be increased by addressing the issues discussed previously mentioned. Nonetheless, the conclusions drawn based on the CFD results remain unaffected by the mentioned error sources.

In vitro\(^{16} \) and in vivo\(^{16} \) experiments showed the importance of the positioning of the outlet graft on the hemodynamics and patient recovery. Results of this study are in agreement with the PIV results in the study by Manning and Miller,\(^{18} \) where adverse- and stagnant-flow regions are shown to depend on the bending angle of the outlet graft. Results of the previous CFD studies\(^{20–22} \) showed that a high-speed jet is formed at the exit of the outlet graft, which causes turbulence and a large increase in WSS. Results reported in the current study confirm these findings.

The angle on the coronal plane primarily affects the helicity in the aortic arch and descending aorta. Maximum WSS and irreversibilities in the anastomosis region rise as \( \phi \) decreases. The azimuthal angle has no effect on the entropy generation. If the inclination angle is chosen as \( 90^\circ \), then 3\% of the work given to the cardiovascular system by the LVAD is dissipated in the anastomosis region. This percentage rises to 5\% as the inclination angle decreases to \( 15^\circ \).

**References**


