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Research Article

Distal Placement of an End-to-Side Bypass Graft Anastomosis: A 3D Computational Study

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A three-dimensional (3D) computational fluid dynamics study of shear rates around distal end-to-side anastomoses has been conducted. Three 51% and three 75% cross-sectional area-reduced 6 mm cylinders were modeled each with a bypass cylinder attached at a 30-degree angle at different placements distal to the constriction. Steady, incompressible, Newtonian blood flow was assumed, and the full Reynolds-averaged Navier-Stokes equations, turbulent kinetic energy, and specific dissipation rate equations were solved on a locally structured multiblock mesh with hexahedral elements. Consequently, distal placement of an end-to-side bypass graft anastomosis was found to have an influence on the shear rate magnitudes. For the 75% constriction, closer placements produced lower shear rates near the anastomosis. Hence, there is potential for new plaque formation and graft failure.

1. Introduction

The leading cause of death in the developed world is the cardiovascular disease, atherosclerosis. It is a progressive disease, in which atherosclerotic plaques, consisting of lipids and cholesterol, slowly develop over time to cause a narrowing of the arterial lumen. The narrowed arterial lumen is called a stenosis, which may grow to significantly reduce or completely obstruct the flow of blood. Turbulence and an adverse pressure gradient may be produced by a stenosis, where separated flow regions could exist and extend several diameters downstream in the poststenotic region.

Surgery is often needed to restore blood flow to tissues affected by a critical arterial stenosis, which is around a 70% cross-sectional area reduction. Commonly, stenotic arteries are repaired by vascular surgical procedures that bypass the stenosis with a conduit called a graft. Furthermore, the end-to-side anastomosis is a common technique used in bypass grafting, where anastomosis refers to a connection between two vessels, and in the case of an end-to-side anastomosis, the end of a graft is attached to the side of an artery with sutures. A significant occurrence associated with this major operation is that it frequently requires revision due to graft failure.

Late graft failure, thirty days after operation, frequently occurs because of normal cell proliferation that results in a thickening of the arterial inner wall called intimal hyperplasia [1]. The primary site for graft failure is the distal anastomosis [2, 3], where intimal hyperplasia is thicker along the floor of the host artery under the anastomosis and along the wall just distal to the toe of the anastomosis [4].

Researchers such as Bandyk et al. [5] and Dobrin et al. [6] have reported that intimal hyperplasia is found in regions of low wall shear stress. Furthermore, researchers such as Caro et al. [7, 8] have observed a possible connection between blood flow and arteriosclerosis by way of low wall shear stress, thereby, leading to a hypothesis correlating low wall shear stress and atherogenesis. In spite of this, it should be noted that atherogenesis may not be a direct result of low wall shear stress, but rather low wall shear stress may result in intimal thickening, which may in turn lead to atherogenesis [9].

Previous computational studies of the effect post-stenotic blood flow phenomena has on wall shear stress around a distal end-to-side anastomosis include Bertolotti and Deplano's [10] low Reynolds number work and a low resolution work by Kute and Vorp [11]. Further investigation is warranted due to the substantial research, indicated previously, that has

shown the proliferation of intimal hyperplasia, which causes late distal graft failure, over a wide range of flow Reynolds numbers, in regions of low wall shear stress. Moreover, if such a relationship between distal placement and wall shear stress is determined, then to some degree vascular surgeons can control the influence of wall shear stress around the anastomosis, which may lead to a prolongation of bypass graft functionality. Accordingly, the present study is concerned with determining how distal placement of an end-to-side bypass graft anastomosis affects wall shear rates.

The following sections describe briefly governing flow equations and computational methods. Then computational results are presented followed by a summary and conclusion of the present study's main findings.

2. Mathematical Formulation

2.1. Flow Equations. Reynolds averaging was used to time-average the instantaneous full Navier-Stokes equations to produce Reynolds-averaged equations of fluid motion, which are better suited to predict the velocity field of a turbulent flow. Assuming the flow is steady and the gravity is negligible, these equations in Cartesian tensor notation are

$$\frac{\partial U_i}{\partial x_i} = 0, \quad (1)$$

$$U_j \frac{\partial U_i}{\partial x_j} = \nu \frac{\partial^2 U_i}{\partial x_j \partial x_j} - \frac{1}{\rho} \frac{\partial P}{\partial x_i} + \frac{\partial \tau_{ij}}{\partial x_j}, \quad (2)$$

where $x_i = (x_1, x_2, x_3)$ represent the Cartesian coordinates, $U_i = (U_1, U_2, U_3)$ represent the Cartesian time-averaged velocity components, P represents the time-averaged pressure, and ν represents the kinematic viscosity, which is a ratio of absolute viscosity to density.

The Reynolds-stress tensor in (2) has introduced six more unknowns in addition to the four unknowns already present, but has not introduced any additional equations. Thus, there is a closure problem. A low Reynolds number form of the $k - \omega$ turbulence model [12] was used to achieve closure. The Boussinesq approach is incorporated relating the Reynolds stresses to the mean velocity gradients. This is represented as

$$\tau_{ij} = \nu_t \left(\frac{\partial U_i}{\partial x_j} + \frac{\partial U_j}{\partial x_i} \right), \quad (3)$$

where ν_t is the turbulent kinematic viscosity and is assumed to be an isotropic scalar quantity. It is expressed as

$$\nu_t = \alpha^* \frac{k}{\omega}, \quad (4)$$

where α^* is the low Reynolds number correction term, which is defined as

$$\alpha^* = \alpha_\infty^* \left(\frac{\alpha_0^* + \text{Re}_t / R_k}{1 + \text{Re}_t R_k} \right). \quad (5)$$

The transport equation for turbulent kinetic energy, k , in Cartesian tensor notation is

$$U_i \frac{\partial k}{\partial x_i} = \frac{\partial}{\partial x_j} \left(\Gamma_k \frac{\partial k}{\partial x_j} \right) + G_k - Y_k. \quad (6)$$

The transport equation for specific dissipation rate, ω , in Cartesian tensor notation is

$$U_i \frac{\partial \omega}{\partial x_i} = \frac{\partial}{\partial x_j} \left(\Gamma_\omega \frac{\partial \omega}{\partial x_j} \right) + G_\omega - Y_\omega. \quad (7)$$

2.2. Computational Details. FLUENT 6.0 is a robust commercial computational fluid dynamics (CFDs) software package that includes the programs GAMBIT 2.0, TGrid 3.4, and FLUENT (the solver). These programs were used in a process to create, to solve, and to interpret the computational tasks of this paper.

During the preprocessing stage, the computational model's coordinate system, geometry, mesh, and boundary conditions were created with GAMBIT, and the quality of the mesh was checked with TGrid. The geometry was based on an ideal femoral artery and bypass graft with 6 mm open diameters. The stenoses studied were 2D in length and smoothly constricted to either a 51% or a 75% area reduction at the throat. Both stenoses resembled an hour-glass shape and were created using cosine equations. In addition, fully developed laminar flow was chosen to flow into the stenosis and into the artery from the graft. Furthermore, the post-stenotic length was chosen to reestablish fully developed laminar flow. Lastly, the anastomotic junction angle was 30 degrees, which is taught to vascular surgeons as an optimal angle (or less) for graft attachment [13].

Six different geometries are presented for this study: three 51% and three 75% arterial models each with a different bypass graft attachment site. Table 1 summarizes the distal bypass graft attachment sites distal to the arterial stenosis, where L_G is the length measured from the throat of the stenosis to the toe of the graft. For case one, the graft was placed in the separation region, and for case two, the graft was placed in the reattachment region of separated flow. Both were also subjected to post-stenotic turbulence. For case three, the graft was placed far downstream to allow a confluence of fully developed laminar flow from the artery and graft. A typical 3D mesh generated in this study is shown in Figure 1, which corresponds to case one.

In addition, a locally structured multiblock mesh was chosen for the present study. A grid sensitivity study was performed on simulations of flow configurations, which correspond to experiments of Ahmed and Giddens [14] and Keynton et al. [15]. Unstructured meshes generated with tetrahedral elements were compared to structured multiblock meshes with hexahedral elements. Mesh quality improved in the latter, with skewness factors going down from 0.3 to below 0.1. Corresponding results were also more accurate. Some mesh sensitivity results are shown in Figure 3.

Geometry of an artery with one end fully occluded and with a graft end-to-side anastomosis at a 30 degree angle was created and meshed. The meshes included: (1) a boundary layer mesh (BL), where cells were clustered toward the walls; (2) a longitudinal mesh (Long), where cells were clustered toward the junction along the x -axis; (3) a refined mesh; and (4) a coarse mesh. Shear rates calculated at the outer wall were normalized with the corresponding shear rate of the graft indicated as "b" in Figure 2. Figure 3 shows the outer

TABLE 1: Summary of graft placements.

	L_G/D	
	51% Stenosis	75% Stenosis
Case 1	3	3
Case 2	~4	~5.7
Case 3	67	67

FIGURE 1: Three-dimensional computational domain and typical mesh of the present study.

wall normalized computational shear rates compared with Keynton et al.'s [15] experiment.

Some additional benefits of using a multi-block mesh include full control of edge mesh density, production of minimally skewed 3D elements for complicated geometries, and accurate and efficient computations.

Conversely, multi-block mesh generation can be much more time consuming than unstructured mesh generation. For instance, the present study's multi-block meshes involved creating and meshing separate blocks that fit together to form the global meshed geometry. After geometry creation, the individual blocks were meshed with a mapping concept that generated eight node hexahedral elements. Moreover, node density was controlled by an interval count spacing function that created $i + 1$ uniformly spaced nodes on an edge, where i is a user defined interval ($i = 0, 1, 2 \dots$).

The solving stage involved the use of the FLUENT solver to define the model's scale, fluid properties, flow physics, solver, and initial conditions. In addition, numerical schemes, such as SIMPLE and second-order upwind-difference approximation, were defined to control the stability and accuracy of the solution of the governing nonlinear partial differential equations. For each model, the mesh files were imported into a double-precision solver. Subsequently, the mesh was checked and then scaled to meters. Then the

FIGURE 2: Axial Measurements Reference Schematic.

FIGURE 3: Normalized Shear Rate at the Outer Wall at $Re = 100$.

steady, segregated solver was chosen along with the default algebraic multigrid solver. In addition, the low Reynolds number formed of the standard $k-\omega$ turbulence model, with default model constants, was selected to predict turbulent flow. Furthermore, Newtonian blood was defined with $\mu = 0.0035 \text{ kg/ms}$ and $\rho = 1060 \text{ kg/m}^3$. The solution was then initialized using the velocity inlet boundary condition, where the Reynolds number was 1,100. Note: first-order accuracy was utilized initially to aid in convergence, and then the models were solved with second-order accuracy with the residual monitor set to 10^{-5} . Experience showed that such residual level was adequate for convergence of the solution in most cases. Furthermore, convergence was assumed when residuals were stable and not decreasing after approximately 1,000 iterations. Under-relaxation factors for pressure, momentum, turbulent kinetic energy, and specific dissipation rate were adjusted as needed. Computations were performed on a single workstation with 2.5 GHz CPU, and typical results were obtained for each case in 10 to 30 hours, depending on the mesh density and the Reynolds number.

Moreover, flow visualization and quantitative analysis of results obtained from solving were carried out during postprocessing and are described next.

3. Results

3.1. Flow Visualization

(a) *51% Stenosis.* The flow field around the distal bypass graft anastomosis (DBGA) placed just distal to the 51% stenosis in a region of separated flow is shown in Figure 4(a). Flow visualization indicates that the flow's velocity increases as it passes through the stenosis and accelerates again just before the anastomotic toe. Moreover, the separation region at the inner wall is truncated by the graft flow. Also, the inner wall separation region appears to be thicker than the outer wall separation region. In addition, the graft flow's momentum causes the arterial flow to skew toward the outer wall under the toe region. Also, graft flow along the outer wall near the heel appears to be detached from the core flow.

For case two, the DBGA was placed in the reattachment region of the separated flow Figure 4(b). Flow visualization shows the flow accelerating through the stenosis followed by a deceleration, where it accelerates once again just before the toe region. Also, the separation region at the inner wall just distal to the stenosis is truncated, and the arterial flow is skewed toward the outer wall by the graft flow beginning under the anastomosis. In addition, flow detachment from the graft outer wall near the anastomotic heel can be seen; however, the graft flow detachment is more conspicuous than for case one.

For case three, placing the DBGA far downstream subjected the region around the anastomosis to fully developed laminar flow and still allowed the influence of the upstream stenosis to be a factor in flow calculations Figure 4(c). Furthermore, the post-stenotic flow separation and turbulence do not directly affect the region around the DBGA, as with the previous two cases. Comparing case three with cases one and two shows that its velocity magnitude is overall slightly less. Similarly, though, the flow from the graft skews the arterial flow toward the outer wall. In addition, flow detachment from the graft outer wall near the anastomotic heel appears to occur at around the same place as for case two; however, graft streamlines are not skewed as much toward the toe.

(b) *75% Stenosis.* The flow field around the DBGA placed in a separation region just distal to the 75% stenosis is shown in Figure 5(a). Similar to Figure 4(a), the separation region at the inner wall is truncated by the graft flow; however, the inner wall separation region is thicker, where its distance from the inner wall is slightly past the inner wall of the throat of the stenosis. Also, flow from the graft is seen causing the arterial flow to skew toward the outer wall in Figure 5(a); however, the impinging graft flow does not cause a noticeable deceleration of the arterial flow under the anastomosis, as was seen in Figure 4(a). Moreover, the graft flow in Figure 5(a) does not appear to detach from the graft outer wall near the heel, which appears to have happened in Figure 4(a).

For case two, the DBGA was placed over the reattachment point of separated flow Figure 5(b). The decrease in cross-sectional area has increased the extent of the separation region. Thus, placement of the DBGA is farther from the stenosis than case two with a 51% stenosis, and as a result, a fairly developed recirculating zone at the post-stenotic inner wall is evident. Also, Figure 5(b) shows a slight increase in overall velocity magnitude from the previous case. Particular to this placement, a confluence of the graft flow and the arterial flow has formed a thin core distal to the anastomotic toe; whereas, for the other placements, the graft flow has noticeably skewed the arterial flow toward the outer wall leaving a thicker higher velocity inner core confluence. Also, Figure 5(b) shows flow detachment from the graft outer wall near the heel.

For case three, the DBGA was placed far downstream in fully developed laminar flow Figure 5(c). Although it is not apparent from the contour coloring, the velocity magnitude of the confluence of arterial flow and graft flow for both Figures 4(c) and 5(c) is similar. In addition, similar flow phenomena are observed, which includes skewing of the arterial flow toward the outer wall caused by the impinging graft flow and flow detachment from the graft outer wall near the anastomotic heel.

3.2. *Shear Rates.* Velocity gradients were measured at the inner wall and outer wall for all six cases. Figure 2, for example, shows how shear rates were measured for computational cases simulating Keynton et al.'s [15] experiment. For the present computational cases, the toe was considered the zero axial reference point for both inner wall and outer wall shear rate measurements. Inner wall measurements were made in 0.5D increments from the toe of the graft to 3D. For the outer wall, measurements were taken in 0.5D increments from -1.5D to 3D. In addition, velocity gradients were normalized with velocity gradients measured at 5D before the throat of the stenosis, where "a" represents the shear rate at the inner wall and "b" represents the shear rate at the outer wall.

The normalized velocity gradients at the inner wall and at the outer wall for the six cases are shown in Figures 6 and 7, respectively. Figure 6 indicates that case three (75%) produced the overall highest inner wall shear rates, and out of the 51% cases studied, case three also showed the highest shear rates. Beginning at the anastomotic toe, shear rates for the 51% cases, as well as the 75% cases, did not differ much from each other. However, further downstream the 75% cases differed significantly. For instance, case one (DBGA placement in a region of separated flow) differed from case three (DBGA placement far downstream) by approximately one normalized magnitude.

For the 51% cases measured beyond the toe, the differentiation among the cases was not as large. For instance, shear rate measurements for case one and case two (DBGA placement in a reattachment region of separated flow) insignificantly differed. Moreover, they differed at most by approximately 0.4 of a normalized magnitude from case three. Interestingly, as axial distance increased along the inner wall, case one (75%) had a normalized shear rate value

FIGURE 4: Velocity magnitude contours with streamlines of the 51% stenosis cases.

similar to cases one and two (51%), and at $L_{iw}/D = 2$ and 2.5, it was lower than these other two cases.

Figure 7 also shows case three (75%) as having the highest overall shear rates, and out of the 51% cases, case three also showed the highest shear rates. For outer wall measurements, cases one and two for both the 51% and 75% stenoses did not vary in magnitude from each other significantly. In addition, before the anastomotic toe, cases one and two (75%) differed on average by approximately two normalized magnitudes from case three, and after the toe, they differed from case three on average by approximately 1.5 normalized magnitudes. Moreover, before the anastomotic toe, normalized shear rates for cases one and two (51%) on average differed by approximately one magnitude, and after the toe, cases one and two (51%) on average differed by approximately 0.6 normalized magnitudes. Interestingly, the highest and lowest outer wall normalized shear rates measured throughout the range of L_{ow}/D studied are observed for 75% stenosis cases.

4. Summary and Conclusion

In summary, the leading cause of death in the developed world is the cardiovascular disease, atherosclerosis. This disease stifles blood flow leading to symptoms that require

surgical intervention. The femoral artery is a common site requiring a bypass graft operation usually involving the end-to-side anastomosis technique. However, late graft failure is likely to occur due to intimal hyperplasia and requires surgical revision. Although the present understanding of the development of atherosclerosis and intimal hyperplasia is limited, hemodynamic theories of vascular pathogenesis offers a promising start into understanding the initiation and proliferation mechanisms for these types of diseases that cause bypass graft failure.

Therefore, different graft attachment locations were studied using a 51% and a 75% stenotic arterial model. Steady, incompressible, Newtonian flow was assumed and simulated on a locally structured multiblock mesh with hexahedral elements. Furthermore, the finite-volume method and a pressure correction scheme, SIMPLE, were used in addition to implicit Gauss-Seidel iteration to solve the full Navier-Stokes equations, turbulent kinetic energy, and specific dissipation rate equations with second-order accuracy. A preliminary study was conducted and confirmed acceptable prediction accuracy of the present computational code and technique. Consequently, the highest shear rates were recorded for the bypass graft anastomosis placed farthest downstream from the stenosis for both the 51% and the 75% stenotic models. Moreover, the lowest inner wall normalized shear rate was

FIGURE 5: Velocity magnitude contours with streamlines of the 75% stenosis cases.

FIGURE 6: Comparison of axial variation of inner wall shear rates for all DBGA cases.

FIGURE 7: Comparison of axial variation of outer wall shear rates for all DBGA cases.

recorded for case one (51%), and the lowest outer wall normalized shear rate was recorded for case two (75%). From Figures 6 and 7, the difference in normalized shear rate magnitudes among the cases indicates that distal placement

of an end-to-side bypass graft anastomosis influences shear rate magnitudes around the distal anastomosis, and with further study, an optimal placement could be determined that possibly extends patent blood flow to afflicted tissues.

As a final note, it is not clear exactly what effect these values of shear rates will have on an *in vivo* distal bypass graft anastomosis; however, previous studies indicate a connection between low wall shear rates and intimal hyperplasia development. The present computational study found the lowest wall shear rates occurring around the DBGAs close to the stenosis for both 51% and 75% cases, where the 75% stenosis on average produced lower shear rates. Therefore, it appears that placement distance from the stenosis and the degree of the stenosis influence shear rates around the distal bypass graft anastomosis.

Nomenclature

D : Diameter
 G : Production term
 i : User-defined interval
 k : Turbulent kinetic energy
 L : Length
 n : Normal to streamline
 P : Time-averaged static pressure
 R_k : Turbulence model constant
 Re_t : Turbulence model term
 u_s : Velocity along streamline
 U : Time-averaged velocity component
 x : Cartesian coordinate
 Y : Dissipation term.

Greek Letters

α^* : Low Reynolds number correction term
 α_0^* : Turbulence model constant
 α_{∞}^* : Turbulence model constant
 Γ : Effective diffusivity
 ν : Kinematic viscosity
 ρ : Density
 τ_{ij} : Reynolds stress tensor
 ω : Specific dissipation rate.

Subscripts

G : Graft
 i, j : Einstein notation (counting index)
 iw : Inner wall
 k : Turbulent kinetic energy
 ow : Outer wall
 t : Turbulent
 ω : Specific dissipation rate.

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