Non-Newtonian Hemodynamics and Shear Stress Distribution in Three Dimensional Model of Healthy and Stented Coronary Artery Bifurcation

Marjan Molavi Zarandi [*, 1], Rosaire Mongrain [1] and Olivier F. Bertrand[2]
[1] Department of Mechanical Engineering, McGill University, [2] Faculty of Medicine, Laval University
[*]Corresponding author: Department of Mechanical Engineering, McGill University, Macdonald Engineering Bui, 817 Sherbrooke Street West, Montreal, Quebec, H3A 2K6 Canada, marjan.molavizarandi@mail.mcgill.ca

Abstract: Stent implantation changes the arterial blood flow patterns and consequently changes the magnitude and distribution of Wall Shear Stress (WSS). Knowledge of local blood flow hemodynamics and shear forces are important for understanding the performance of stents. In certain regimes within the stented region (low shear rate regimes) blood should be treated as a non-Newtonian fluid. In this paper, a three-dimensional model of the coronary artery bifurcation is developed and blood flow hemodynamics and WSS distribution in both healthy and stented coronary artery bifurcation are analyzed using COMSOL Multiphysics. Our simulation shows that the presence of stent within the bifurcation induces local disturbance in the flow field and consequently produces regions of low and non-uniform shear stress in the main and daughter vessel.

Keywords: Wall Shear Stress, Stent, non-Newtonian, Hemodynamics, Bifurcation.

1. Introduction

Arterial diseases like atherosclerosis are the leading causes of death in the industrial world. Atherosclerosis is a condition in which fatty material collects along the arterial. It reduces arterial lumen size through plaque formation and growth [1]. Pathologic examination of coronary arteries reveals that atherosclerosis lesions occur at particular locations of the arterial tree including bifurcations and curvatures [2]. Local factors, such as hemodynamic forces and wall shear stress, play a major role in the localization and development of atherosclerosis [3- 4].

Open heart surgery was the main method for the treatment of cardiovascular diseases until endovascular therapies were developed. In the early 1980s, stents, which could be expanded into the stenotic artery and restore the normal blood flow in the artery, were presented.

Stent implantation changes the arterial blood flow patterns and subsequently changes the magnitude and distribution of shear stress. Peacock *et al.* demonstrated measurable flow disturbances due to stent implantation using an in vitro pulse duplication system [5]. Both in vitro and in vivo studies have revealed that stent structure influences global and local flow patterns [6, 7]. Studies of stent-induced changes in artery wall flow patterns, focused on the near-wall flow patterns, revealed very low magnitude of shear rate between the stent struts [8].

Bifurcation lesions have always been a major challenge for placement of stents for treatment of stenosis. Conventional stents are mostly used in clinical practice since there is no specific commercially available stent dedicated for treating bifurcations. In this context, what was previously assessed for standard stents should be re-assessed for stenting bifurcations.

Investigation of blood flow hemodynamics and shear forces are of great importance in understanding the regions of disease formation, development and predicting the performance of cardiovascular technologies such as stents. The flow conditions and particularly shear stress distribution near wall in stented coronary artery are expected to be different compared to the healthy bifurcation. However, the study of blood flow hemodynamics requires an in depth understanding of the physics and assumptions toward a realistic physiological modeling.

This work will address these issues using an approach incorporating the physiological conditions. An anatomical structure is replicated and three-dimensional Computational Fluid Dynamic (CFD) analysis is carried out to analyze the hemodynamic changes induced by endovascular stents in the main and daughter vessel of coronary artery bifurcation considering non-Newtonian flow model.

2. Material and Methods

A Three-dimensional model of the stented bifurcation was developed and computational fluid dynamics is employed using COMSOL to determine the corresponding velocity field and shear stress distribution in the arterial wall.
The geometrical model of the bifurcation consisted of the left main coronary artery and a 45º bifurcation of two daughter vessels. This model is similar to the bifurcation between the left anterior descending artery and coronary artery. The parent vessel diameter was 4 *mm* and daughter vessels diameters were 3.2 *mm* and 3.1 *mm* respectively [9].

The dimensions which were adopted for this model are 5 struts, 0.15 *mm* in diameter, and located 0.7 *mm* center to center apart (Fig. 1). These stent dimensions are typical of common coronary stents [10].

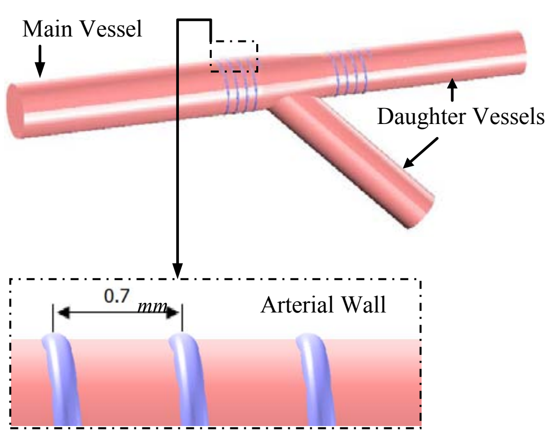


Figure1. Geometrical model of stented coronary artery bifurcation.

For the study of blood flow in the arteries, we assume that blood can be represented by an incompressible fluid which is governed by the momentum equations:

$$\rho(u.\nabla u) = -\nabla p + \nabla.\tau \qquad (1)$$

and the continuity equation:

$$\nabla.u = 0 \qquad (2)$$

where, ρ denotes the density of the fluid ($kg\ m^{-3}$), u the velocity vector (*m s-1*), p the pressure (*Pa*) and τ the stress tensor which is dependent on the viscosity and shear rate. The Carreau model approaches the constant Newtonian value at high shear rates and provides adequate information to model blood behaviour [11].

The non-Newtonian blood properties in this model are blood viscosity at infinite shear rate μ_∞ =0.035 *mPa.s*, blood viscosities at zero shear rates μ_0=0.56 *mPa.s*, λ=3.313 *s* and n=0.3568 [11]. Blood density is ρ =1060. In the Carreau model, the viscosity is modeled by the equation:

$$\eta = \mu_\infty + (\mu_0 - \mu_\infty)[1+(\lambda\dot{\gamma})^2]^{\frac{(n-1)}{2}} \qquad (3)$$

Here η is the effective apparent blood viscosity, μ_∞ and μ_0 are the blood viscosities at infinite and zero shear rates (*Pa. s*) respectively, is the shear rate (s^{-1}); λ is a time constant, and n is power law index which is a dimensionless parameter determined with experimental fit [12]. In the Carreau model, the viscosity is dependent on the shear rate ($\dot{\gamma}$), which for three dimensions is defined according to equation:

$$\dot{\gamma} = [2\{(\frac{\partial u}{\partial x})^2 + (\frac{\partial v}{\partial y})^2 + (\frac{\partial w}{\partial z})^2\} + (\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x})^2 + (\frac{\partial u}{\partial z} + \frac{\partial w}{\partial x})^2 + (\frac{\partial v}{\partial z} + \frac{\partial w}{\partial y})^2]^{\frac{1}{2}} \qquad (4)$$

where u, v and w are the x, y and z components of the velocity vector, respectively [12].

To solve the governing equations, a set of boundary conditions is required. In this analysis, the mean velocity of 28 *cm/s* was considered for a coronary artery of 4 *mm* in diameter [10]. The flow was considered laminar and fully developed throughout the study section. At the walls, the velocity obeyed the no-slip condition. At the outlet of the daughter vessels, a stress-free condition was assumed as the boundary conditions.

3. Results

Numerical analysis of blood flow in healthy and stented coronary artery bifurcation was carried out for non-Newtonian flow model. The problem was solved based upon the steady laminar flow of a homogeneous, incompressible, non-Newtonian fluid through a bifurcated channel with rigid walls. The velocity field in the main vessel and daughter vessels of bifurcation is shown in figure 2.

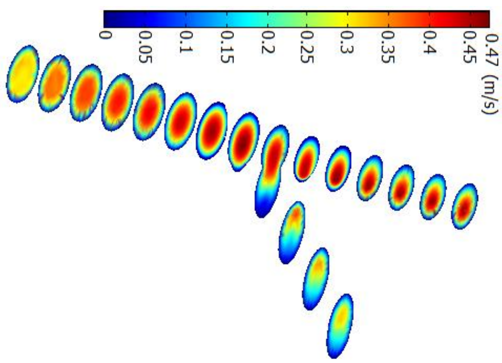


Figure 2. Velocity field in main vessel and daughter vessels of coronary artery bifurcation.

The velocity profile shows that the velocity deflects toward the lower wall immediately downstream of the bifurcation.

Wall shear stress which is the magnitude of the tangential shear forces acting on the wall by the fluid is determined by the wall shear rate (gradient of velocity at the wall) multiplied by the viscosity of the fluid. Therefore, the gradient of velocity creates significant changes in the wall shear stress on the upper and lower walls of main and daughter vessels. Distribution of WSS in the healthy artery is shown in Figure 3.

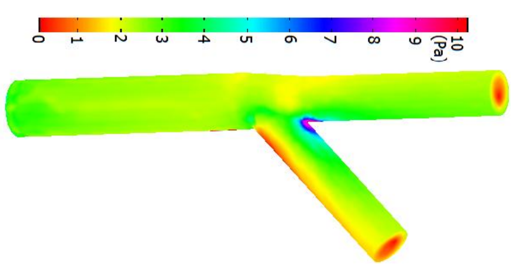
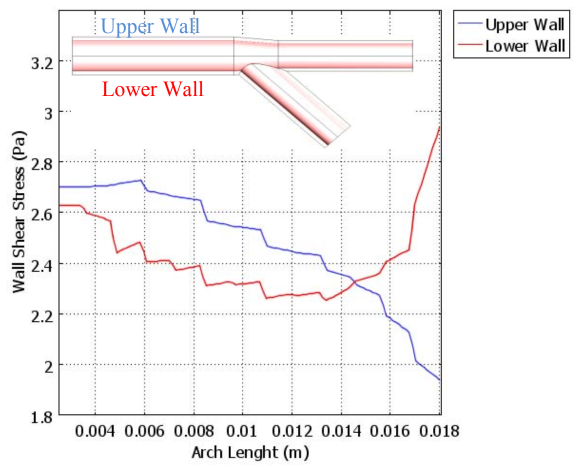
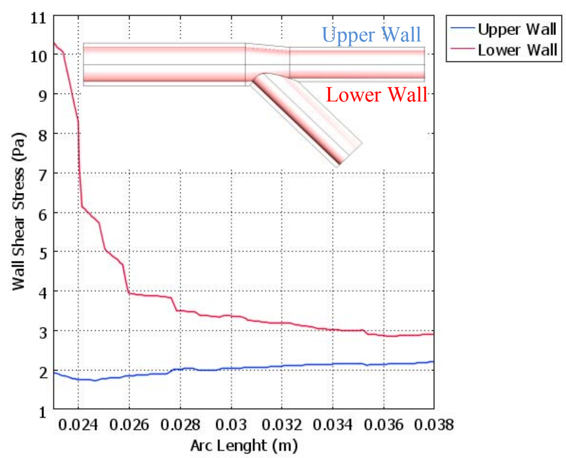


Figure 3. Distributions of WSS in the healthy coronary artery bifurcation.

Values of WSS were obtained for the upper and lower walls of main vessel and daughter vessel by looking at a plane section through the longitudinal axis of the bifurcation. Figure 4 shows the WSS values in the main and daughter vessels before stenting bifurcation.



(a)



(b)

Figure 4. Distributions of WSS on the upper wall and lower wall of main vessel (a) and daughter vessel (b) before stenting bifurcation.

Following the stent deployment in the arterial wall, the variations in blood velocities created significant changes in the wall shear stress. Figure 5 shows the distribution of wall shear stress in the main and daughter vessels after stent deployment in the bifurcation.

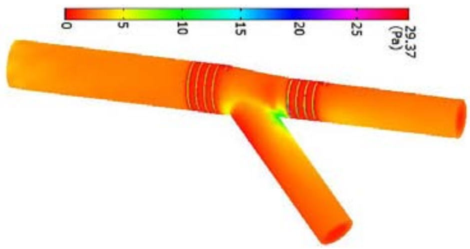
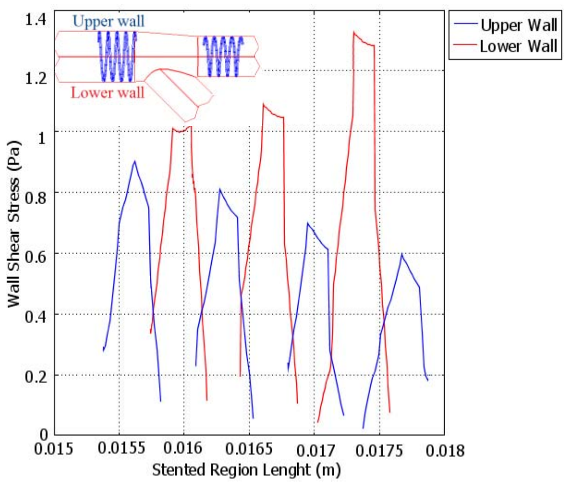
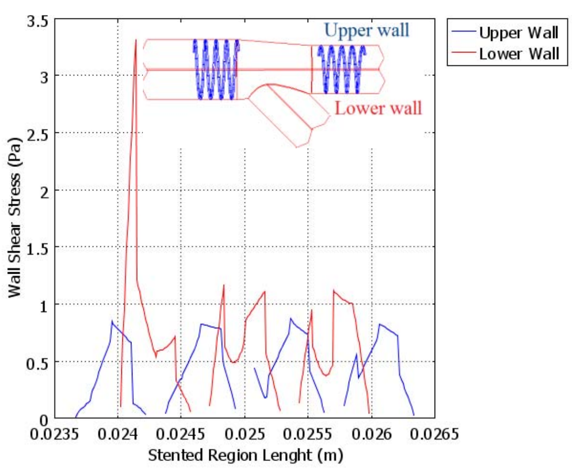


Figure 5. Wall Shear Stress after stenting bifurcation.

Wall shear stress distribution at the upper and lower walls of main vessel and daughter vessel in the stented segment at the plane section through the longitudinal axis of the bifurcation is shown in Figure 6.



(a)

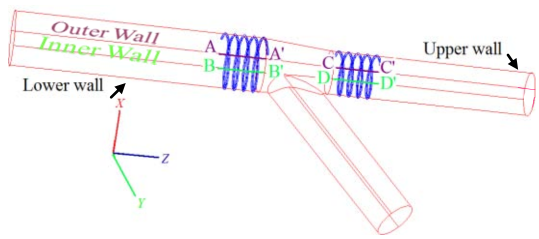


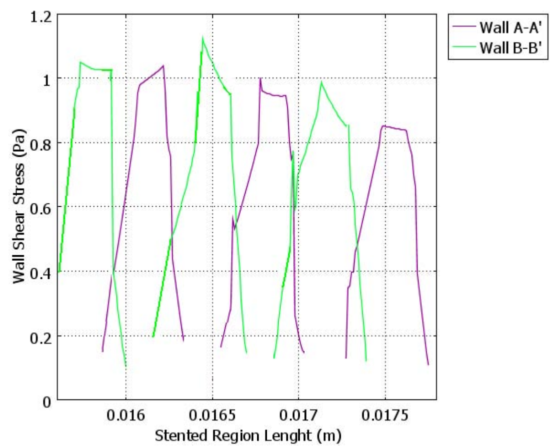
(b)

Figure 6. Wall Shear Stress on the upper and lower walls of main vessel (a) and daughter vessel (b) after stenting bifurcation.

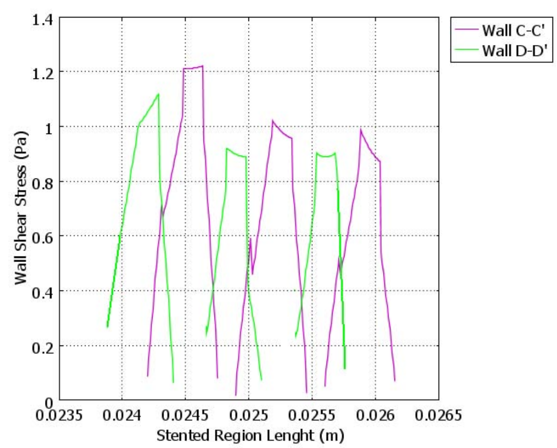
Comparing Figures 4 and 6 shows a significant decrease in the magnitudes of wall shear stress along the arterial wall after stenting bifurcation.

Lateral wall shear stress values of the main vessel and daughter vessel in the stented segment were obtained in the plane section through the lateral axis (*Y-Z*) of the bifurcation as shown in Figure 7.





(a)



(b)

Figure 7. Wall Shear Stress on the lateral walls of main vessel (a) and daughter vessel (b) after stenting.

These results make it possible to compare the WSS distributions for inner and outer walls of the main vessel and daughter vessel in the lateral direction. Although there is a significant decrease in the magnitudes of WSS in the stented bifurcation compare to the healthy artery, WSS values doesn't change remarkably on the inner wall outer wall of bifurcation in the lateral direction.

4. Conclusions

In this study, we used computational fluid dynamic simulations to investigate the effect of stent deployment on the flow patterns and WSS distribution in the stented arterial bifurcation.

Our simulations showed that the presence of stent within the bifurcation induces local disturbance in the flow field and consequently produces different shear stress in the walls of bifurcation. It was shown that the lowest value of WSS belongs to the intra-stent region of the main and daughter vessels and on the upper walls.

It was observed from this study that the higher wall shear stress variation was occurred at the longitudinal direction and on the upper and lower walls of main and daughter vessels in the stented segments. In the lateral direction, there was not a significant difference between WSS values on the inner wall outer wall of stented bifurcation. It means that for the specific type of stented bifurcation presented in this study, different stent design or stent strut design should be considered for upper and lower walls of both main and daughter vessels in order to have appropriate distribution of WSS in to the arterial wall.

5. References

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