THE INFLUENCE OF STENT GEOMETRY AND NON-NEWTONIAN EFFECT ON HEMODYNAMICS IN STENTED CORONORY ARTERY

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Abstract

Stents are tubular structure equipments which deployed to physically reopen stenotic regions of arteries. Although stents are widely used to restore blood flow, they increase the risk of restenosis in the vicinity of stenotic area. It seems that cross sectional geometry of stent has main effect on artery restenosis. In this study, tubular stents with different cross sectional geometries are considered in a coronary artery and Newtonian and non-Newtonian viscosity models are utilized to simulate blood rheology. The results show that streamlined stents prevent low and high wall shear stresses in the vicinity of the stent. Also, these stents decrease the risk of thrombosis and restenosis by eliminating the particle residence time in the region of stented vessel. On the other hand, results indicate that Newtonian model underestimates the wall shear stress within a stented arterial segment, which can lead to an overestimation of the risk of restenosis.

Key words: Stent, Coronary artery, Hemodynamic, Wall shear stress,

1. Introduction

Stents are tubular structures which are used to restore blood flow through stenotic region especially in carotid and coronary arteries. Stents are inserted to the stenotic region after percutaneous transluminal angioplasty. Unfortunately, restenosis, which is the re-narrowing of the lesion after implantation stent, remains a persistent problem [1]. Previous investigations indicated that restenosis for bare metal stents can reach to 20-40% [2]. Although the restenosis for drug eluting stents is about 5% [3, 4] but late thrombosis has been considered to be a long term issue in drug-eluting stents [5].

It has been exhibited that wall shear stress (WSS) and wall shear stress gradient (WSSG) have major impact on restenosis [6, 7]. Mongrain et al [8] showed that areas with low WSS (<1.26 Pa) displayed a significant increase in the thickness of the atherosclerotic plaque and the vessel wall (positive remodeling). Areas with physiologic WSS (1.26-2.69 Pa) did not display significant changes and areas exposed to a high WSS (≥2.7 Pa) displayed positive remodeling of the artery without changes in the atheroma plaque. Also Ku et al [9, 10] reported a strong inverse correlation between low mean wall shear stress and atherosclerotic intimal thickening.

It seems that WSS strongly dependent on stent type and its geometry [11]. There are quite a few studies which investigate the effect of stent geometries such as strut thickness, interstrut spacing strut-connector and strut radius of curvature on local hemodynamic which demonstrated that stent geometry and its subsequent effects on localized fluid dynamics is an important factor on restenosis[12-16]. In previous studies the effect of cross sectional geometry on hemodynamic is not well established so it is vital to assess this effect. On the other hands the vast majority of studies concerning stents [17-20] have considered blood as a Newtonian fluid with a viscosity equal to blood's viscosity at very high shear strain rates (i.e., 3.45 cp). But this assumption recently is in doubt [21].

The present study is dedicated to evaluate the effect of cross sectional geometry of stent on hemodynamic and the restenosis of treated arteries. Besides, the wall shear stress and separation distance are compared for Non-Newtonian and Newtonian models.

2. Materials and Methods

To evaluate the effect of stent geometry on restenosis, six different cross-sectional stent strut geometries with different aspect ratio are used. The width of each struts (w) remain constant equal to 0.2mm while the height (h) changes in each case (Figure 1). Each stent includes six struts which are set along each other with specified distance (i.e., 2mm). Additionally, all cases have length of 19.2mm and 3mm diameter (Figure 2).

An axisymmetric steady state fluid flow solved numerically with control volume approach. The governing equations were solved using a commercial computational fluid dynamics (CFD) software package (Fluent). A no-slip boundary condition is set at all walls, the inlet boundary conditions correspond to a fully developed circular duct profile .Inlet mean velocity assessed base on Reynolds number ($Re=\rho VD/\mu$) of 350 that predominates in stented vessel, at the exit boundary, zero value was specified for the pressure and a symmetry condition was applied to the centerline of the vessel. It also should be noted that vessel is assumed to be rigid and the rigidity assumption is reasonable [22].

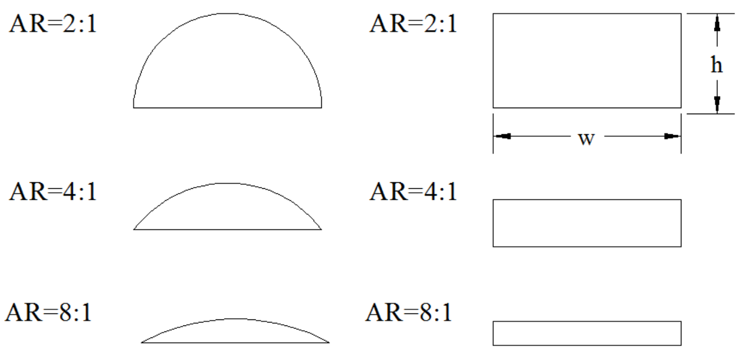

Figure 1: Cross-sectional stent strut geometries with different aspect ratio, AR= width to height

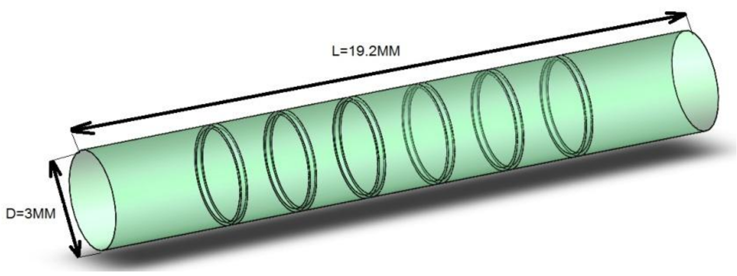

Figure 2: Stented coronary artery with six independent rings representing the stent struts

2.1. Governing Equations

The following mass conservation (1) and momentum conservation (2 and 3) equations were solved using Fluent Computational Fluid Dynamics (CFD) software.

$$\frac{\partial u}{\partial x}+\frac{1}{r}\frac{\partial (rv)}{\partial r}=0 \tag{1}$$

$$\rho\left(\frac{\partial u}{\partial t}+u\frac{\partial u}{\partial x}+v\frac{\partial u}{\partial r}\right)=-\frac{\partial P}{\partial x}+\mu\left[\frac{1}{r}\frac{\partial}{\partial r}\left(r\frac{\partial u}{\partial r}\right)+\frac{\partial^2 u}{\partial x^2}\right] \tag{2}$$

$$\rho\left(\frac{\partial v}{\partial t}+u\frac{\partial v}{\partial x}+v\frac{\partial v}{\partial r}\right)=-\frac{\partial P}{\partial r}+\mu\left[\frac{1}{r}\frac{\partial}{\partial r}\left(r\frac{\partial v}{\partial r}\right)+\frac{\partial^2 v}{\partial x^2}-\frac{v}{r^2}\right] \tag{3}$$

In above equations, u and v are the axial and radial velocity components; $\boldsymbol{\rho}$, μ and P represent blood density, dynamic viscosity, and pressure, respectively. Blood flow is assumed to be pulsatile , axisymmetric, incompressible and laminar with a density of 1060 kg/m3. Dynamic viscosity is assumed 3.45cp for Newtonian model. The Carreau-Yasuda model with the following equation is utilized to simulate Non-Newtonian behavior of blood.

$$\mu=\mu_\infty+(\mu_0-\mu_\infty)(1+(\lambda\gamma)^2)^{\frac{n-1}{2}} \tag{4}$$

Where μ_0 is zero shear rate viscosity, μ_∞ is infinite shear rate viscosity, λ is time constant, γ is shear rate, and n is power index. The Carreau-Yasuda model coefficients for real blood are μ_0 = 56 cP, λ = 3.313, μ_∞= 3.45 cP, n = 0.3568 [23].

The wall shear stress is defined by equation (5) where $\dot{\gamma}$ is the shear strain rate.

$$\tau_w=\mu\left(\frac{\partial u}{\partial r}+\frac{\partial v}{\partial x}\right)=\mu\dot{\gamma} \tag{5}$$

2.1. Mesh Generation

The mesh generated by quadrilateral elements for rectangular strut geometries while quadrilateral and triangular element is used for circular strut geometries. Figure 3 shows the mesh density in area of the third strut. Mesh independence is a necessary condition for the validity of the velocity and wall shear stress. Mesh independence was obtained for a mesh density of 525000 elements for rectangular and 250000 elements for circular strut geometries.

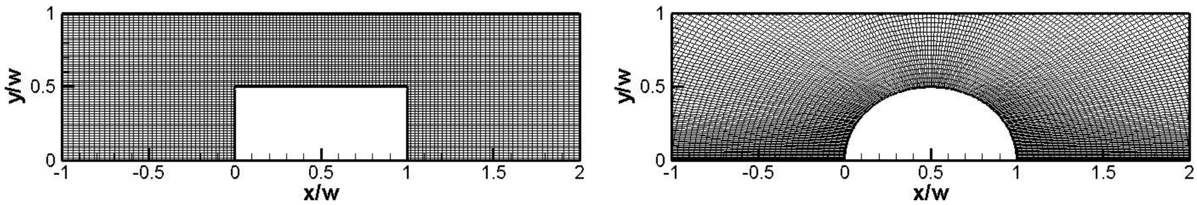

Figure 3: Grid spacing meshes in the vicinity of a 2:1 aspect ratio rectangular and circular stents strut.

3. Results and Discussion

Six different strut geometries for Newtonian and Non-Newtonian models are considered in stented coronary artery to report the following result. For each case of study, pressure field, separation distance, wall shear stress and shear rate are reported, and the effects of Non-Newtonian rheology on the results are studied. All the results in this study belong to the third strut. Besides, axes corresponding to distance are nondimensionalized by w and the location of the leading edge of the third strut is set to zero.

3.1. Pressure field

The nondimensional pressure field has been showed in figure 4. The pressure is non dimensiolized by dividing the static pressure by the dynamic pressure $p^* = p/\frac{\rho\bar{U}}{2}$. Higher pressure is observed on the upstream side of the strut and it is decreased by increasing the x/w. For rectangular struts the maximum of pressure was found in the upstream tip of the strut, but for circular struts this effect just only exists on 2:1 aspect ratio due to its nonstreamlined geometry. Pressure gradient has the greatest value in 2:1 AR rectangular strut and it decreases by decreasing the height of rectangular struts. The lowest pressure gradient was found in 8:1 circular struts.

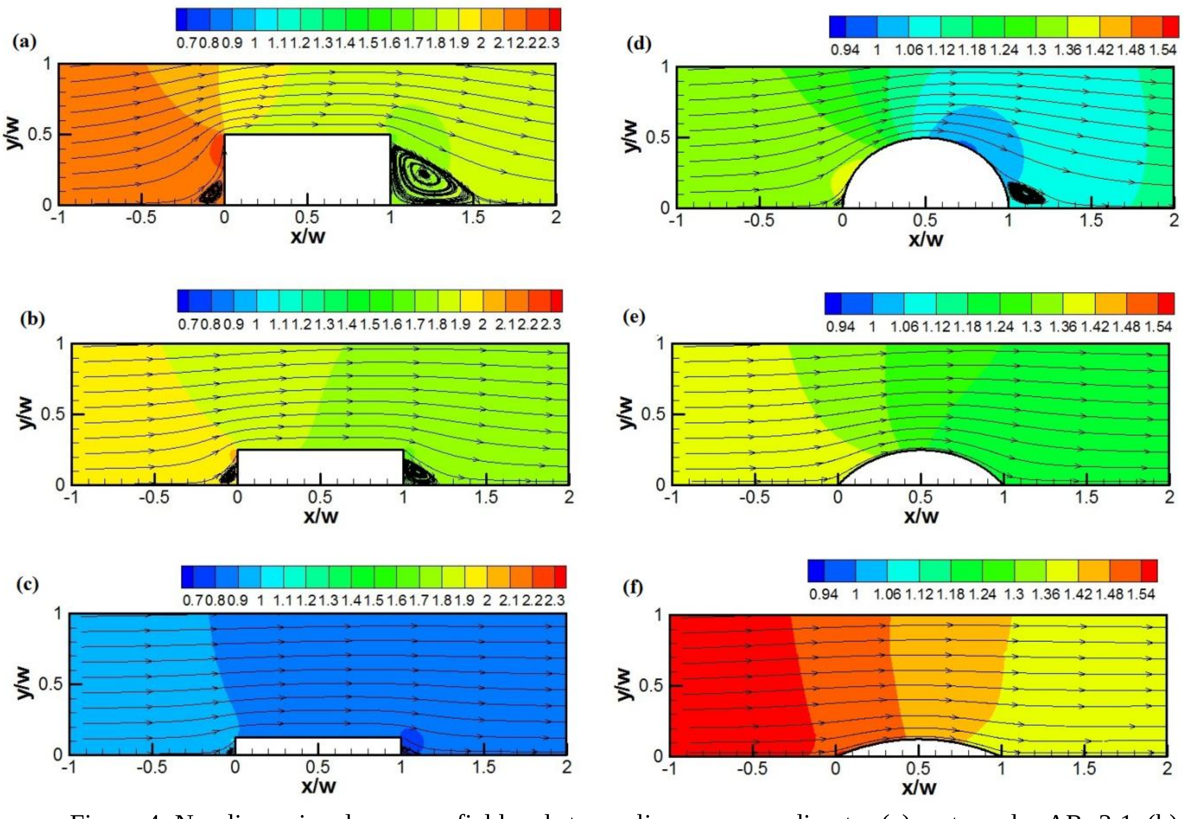

Figure 4: Nondimensional pressure field and stream line corresponding to: (a) rectangular AR=2:1, (b) rectangular AR=4:1, (c) rectangular AR=8:1, (d) circular AR=2:1, (e) circular AR=4:1,and (f) circular AR=8:1 for Non-Newtonian model

3.2. Separation distance and zone

Table 1 shows the separation distance which it nondimensionalized by the width of the strut. Newtonian model exhibits longer upstream and downstream separation length in comparison with Non-Newtonian model, so Newtonian model evaluates the vessel condition worse than its reality. In addition it is found that stream lined geometry such as circular 4:1 and circular 8:1 prevent separation zone in vicinity of the strut, which may leads to lower risk of restenosis. Result comparison indicates that rectangular strut with 2:1 aspect ratio has the longest separation length, which causes accumulation of plaques in the vicinity of strut and increases the rate of restenosis.

Table 1: Nondimensional separation length upstream and downstream

	Newtonian model		non-Newtonian model	
Geometry	Upstream distance	Downstream distance	Upstream distance	Downstream distance
Rectangular 2:1	0.245	0.795	0.190	0.540
Rectangular 4:1	0.155	0.250	0.140	0.225
Rectangular 8:1	0.075	0.090	0.065	0.075
Circular 2:1	0.100	0.420	0.080	0.305

3.3. Wall shear stress

Wall shear stress and shear rate of strain are plotted along the stented region in figures 5 and 6 for Newtonian and Non-Newtonian models, respectively. For different geometries, results show that high value of WSS is observed on the top of the struts. The highest value of WSS for rectangular strut was in the upstream near the strut and for circular struts it was around the middle of the strut. Figure 5 shows that the struts with higher aspect ratio result in lower values for wall shear stress. Generally, it should be noted that nonstreamlined stent struts geometries such as rectangular cross-sectional geometries can reach higher levels of WSS in comparison to streamlined geometries

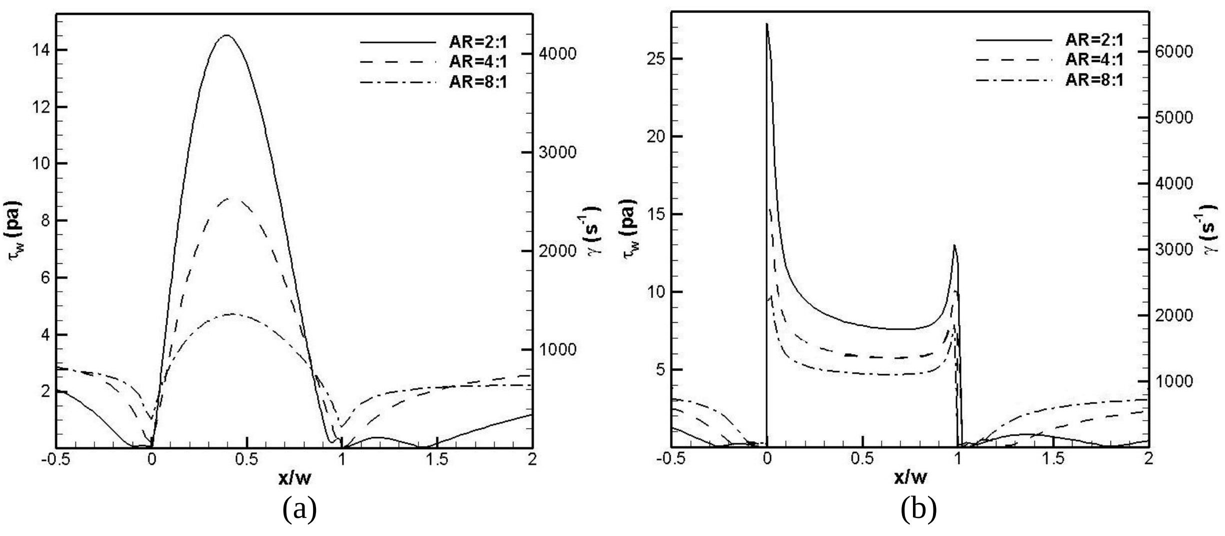

Figure 5: WSS and shear rate distribution for (a) circular and (b) rectangular stent struts for different aspect ratio in Newtonian model

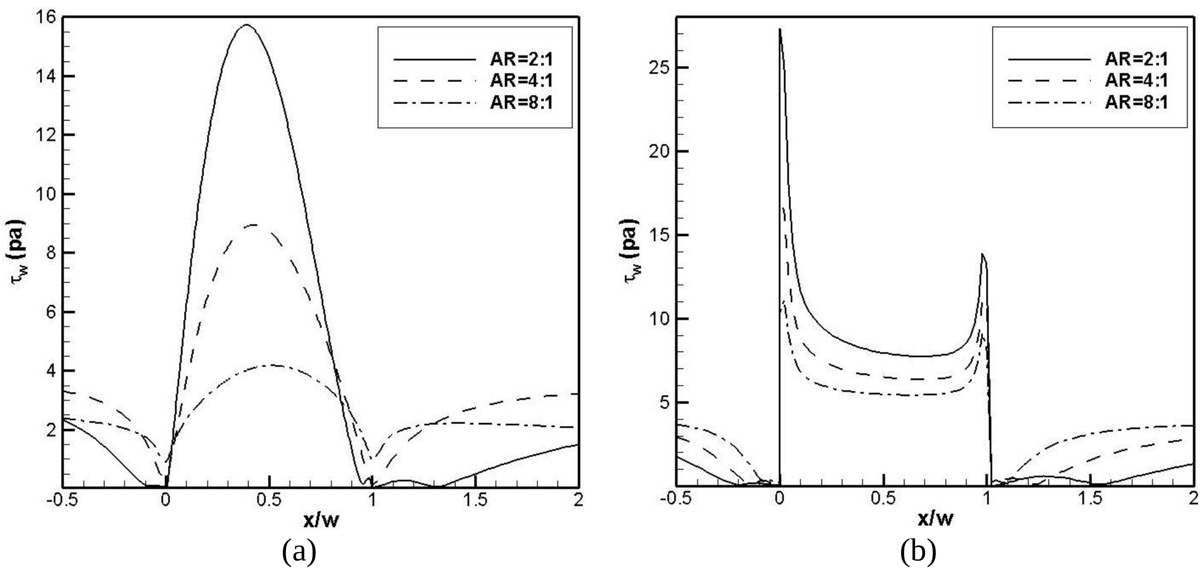

Figure 6: WSS distribution for (a) circular and (b) rectangular stent struts for different aspect ratio in Non-Newtonian model

Comparison of WSS plots for Newtonian and Non-Newtonian models confirmed that they are very similar to each other except in the region where WSS is less than critical value of 1.26 Pa. Results show that on the downstream of the strut, especially in circular and rectangular struts with 2:1 aspect ratio, the Newtonian model exhibits longer region which the WSS is below than its critical value. In Newtonian model the critical distance, which WSS is below than 1.26 Pa, is 5% and 26% more than the Non-Newtonian model for rectangular and circular struts, respectively. It demonstrated that Newtonian model evaluates the condition of the stented vessel more critical than reality which is in agreement with Mejia et al study [21].

4. Conclusions

Present study shows that non-streamlined geometry such as rectangular stent strut can minimize the circulation area by decreasing the height of struts. Rectangular strut experiences very high wall shear stress gradient at the beginning and ending of the strut which can speed up restenosis. On the other hand, streamlined cross section geometry can minimize or eliminates circulation area and prevents high wall shear stress gradient in the vicinity of the strut. Results demonstrated that Newtonian model exhibits larger separation distance and also greater region exposed to low shear stress. It means that Newtonian model shows the condition of stented vessel more critical in compare with Non-Newtonian model.

Reference

[1] R. Erbel, M. Haude, H. W. Hopp, D. Franzen, H. J. Rupprecht, B. Heublein, K. Fischer, P. de Jaegere, P. Serruys, W. Rutsch, and P. Probst, Coronary-artery stenting compared with balloon angioplasty for restenosis after initial balloon angioplasty, Restenosis Stent Study Group. N. Engl, J. Med. 339, 1672–1678 (1998).

[2] A. Kastrati, J. Mehili, J. Dirschinger, J. Pache, K. Ulm, H. Schuhlen, M. Seyfarth, C. Schmitt, R. Blasini, F. J. Neumann, and A. Schomig,, Restenosis after coronary placement of various stent types. American Journal of Cardiology., 87(1), 34-39, (2001).

[3] T. C. Woods and A. R. Marks, Drug-eluting stents, Annual Review of Medicine., 55, 169-178 (2004).

[4] M. A. Castro, C. M. Putman, and J. R. Cebral, Computational fluid dynamics modeling of intracranial aneurysms: Effects of parent artery segmentation on intra-aneurysmal hemodynamics. American Journal of Neuroradiology., 27(8), 1703-1709 (2006).

[5] MH. Wernick, A. Jeremias, and JP. Carrozza, Drug-eluting stents and stent thrombosis: a cause for concern? Coron Artery Dis, 17(8):661–5, (2006).

[6] S.G. Carlier, L. C.A. van Damme, C. P. Blommerde, J. J. Wentzel, G. V. Langehove, S. Verheye, M. M. Kockx, M. W. M. Knappen, G. Cheng, F. Gijsen, D. J. Dunker, N. Stergiopulos, C. J. Slager, P. W. Serruys, and P. Krams, Augmentation of wall shear stress inhibits neointimal hyperplasia after stent implantation Inhibition through reduction of inflammation, Circulation., 107(21), 2741-2746 (2003)

[7] C. Kleinstreuer, S. Hyun, J. R. Buchanan, P. W. Longest, J. P. Archie, G. A. Truskey, Hemodynamic Parameters and Early Intimal Thickening in Branching Blood Vessels, Crit Rev Biomed Eng., 29, 1-64 (2001).

[8] R. Mongrain and J. Rodés-Cabau, Role of shear stress in atherosclerosis and restenosis after coronary stent implantation, Rev. Esp. Cardiol., 59, 1-4, (2006).

[9] D. N. Ku, Blood flow in arteries, Annu. Rev. Fluid. Mech.,29,399–434, (1997).

[10] D. N. Ku, C. K. Zarins, D. P. Giddens, and S. Glagov, Pulsatile flow and atherosclerosis in the human carotid bifurcation: positive correlation between plaque localization and low and oscillating shear stress, Arteriosclerosis., 5:292–302 (1985).

[11] J. F. J. LaDisa, I. Guler, L. E. Olson, D. A. Hettrick, J. R. Kersten, D. C. Warltier, and P. S. Pagel, Three dimensional computational fluid dynamics modeling of alterations in coronary wall shear stress produced by stent implantation, Ann. Biomed. Eng., 31,972–980, (2003).

[12] S. Pant, N. W. Bressloff, A. I. J. Forrester, and N. Curzen, The Influence of Strut-Connectors in Stented Vessels: A Comparision of Plusatile Flow Through Five Coronary Stents, Annals of biomechanical engineering., DOI: 10.1007/s10439-010-9962-0 (2010).

[13] R. Balossinio, F. Gervaso, F. Migliavacca, and G. Dubini, Effects of different stent design on local hemodynamics in stented arteries, Jornal of Biomechanics., 41, 1053-1061 (2008).

[14] I. Faik, R. Mongrain, R. L. Leask, J. Rodes-Cabau, E. Larose, and O. Bertrand, Time-dependent 3D simulation of the hemodynamic in a stented coronary artery, Biomed. Mater., 2,28-37 (2007).

[15] Y. He, N. Duraiswamy, A. O. Frank, and J. E. Moore, Blood flow in stented arteries: a parametric comparison of strut design patterns in three dimensions. J. Biomech. Eng., 127, 637–647 (2005).

[16] A. Kastrati, J. Mehilli, J. Dirschinger, F. Dotzer, H. Schuhlen, F. J. Neumann, M. Fleckenstein, C. Pfafferott, M. Seyfarth, and A. Schomig, Intracoronary stenting and

angiographic results: strut thickness effect on restenosis outcome,(ISAR-STEREO) trial. Circulation., 103, 2816–2821 (2001).

[17] J. M. Jimenez and P. F. Davies, Hemodynamically driven stent Strut design, Annals of Biomechanical engineering, Vol. 37, No.8 ,1483-1494 (2009).

[18] S. G. Carlier, L. C. A. van Damme, C. P. Blommerde, J. J. Wentzel, G. van Langehove, S. Verheye, , M. M. Kockx, M. W. M. Knaapen, C. Cheng., F. Gijsen, D. J. Duncker, N. Stergiopulos, C. J., Slager, P. W. Serruys, and R. Krams, Augmentation of Wall Shear Stress Inhibits Neointimal Hyperplasia After Stent Implantation: Inhibition Through Reduction of Inflammation?, Circulation.,107, 2741–2746 (2003).

[19] J. F.J. LaDisa, L. E. Olson, I. Guler, D. A. Hettrick, J. R Kersten, D. C. Warltier, and P. S. Pagel, ,Circumferential Vascular Deformation After Stent Implantation Alters Wall Shear Stress Evaluated Using Time- Dependent 3d Computational Fluid Dynamics Models, J. Appl. Physiol., 98, 947–957 (2005).

[20] N. Duraiswamy, R. T. Schoephoerster, M. R. Moreno, and J. E. Moore, Stented Artery Flow Patterns and Their Effects on the Artery Wall, Annu. Rev. Fluid Mech., 39, 357–382 (2007).

[21] J. Mejia, R. Mongrain, O. F. Bertrand, Accurate prediction of Wall shear Stress in a stented Artery: Newtonian Versus Non-Newtonian Models, Journal of biomechanical engineering, Vol., 133/074501-1 (2011).

[22] D. N. Ku, Blood flow in arteries, Annu. Rev. Fluid. Mech., 29, 399–434 (1997).

[23] YI. Cho, KR. Kensey, Effects of the non-Newtonian viscosity of blood on flows in a diseased arterial vessel. Part 1: Steady flows, Biorheology, 28(2-4), 241-62 (1991).