CFD comparative study of a flat versus a cylindrical stent configuration for PIV investigations of the prosthetic influence on local hemodynamics.

G. Drapeau[1], R. Mongrain[*,2], Juan Mejia[2], J. Rodes-Cabau MD[3] and O.F. Bertrand MD[3]
1)Biomedical Engineering Departement and 2)Mechanical Engineering Departement
McGill University, Montreal, QC, Canada
3)Laval Hospital Research Center, Quebec, Canada.
[*] Corresponding author: rosaire.mongrain@mcgill.ca, (514) 398-1576

Abstract

Mechanical stresses and flow dynamics alteration in a stented artery region are known to stimulate intimal thickening and increase the risk of restenosis, the closure of a revascularized artery. Particle imaging velocimetry flow visualization technique (PIV) can be useful in characterizing the local flow dynamics around stents. However, usual cylindrical stent models present some visualization and optical difficulties when studying fluid dynamics using PIV. A flat stent model has two main advantages over the usual cylindrical models. First, planar stent models are cheaper to manufacture than cylindrical ones. Secondly, it is easier with a flat model to get data acquisition planes cutting through the model without inducing diffraction errors due to the cylindrical model's curvature. The wall shear stress distribution, which is believed to be of primary importance in stented artery restenosis, should be comparable in both the flat and cylindrical stent models for the regions of interest. In order to study the feasibility of using a flat version of the stent model for PIV flow investigation the shear stress distribution computed in the CFD analysis will be compared between the flat and the cylindrical stent configurations. It will be shown that for a physiological pulsatile flow the flat model yields results in the shear strain rate that are comparable to the cylindrical model. A cheaper and less diffraction error prone validated flat model would be advantageous when several stent design have to be studied and optimized.

Keywords: Stent, restenosis, Particle imaging velocimetry, pulsatile flow, CFD.

1. Introduction

Stents are frequently used, during angioplasty, as a method to reduce the risk of restenosis (i.e. the re-narrowing of the artery). Restenosis, however, still occurs at a rate of about 20% in stented vessels [1, 2]. Local hemodynamics, particularly the wall shear stress, around the struts of a stented artery region and the risk of restenosis are believed to be correlated [3]. PIV, particle imaging velocimetry, has been used successfully in studying hemodynamics around the stent struts in scaled-up transparent stent models [4]. Roughly put, the PIV technique consists in tracking particles that reflect light coming from a thin laser sheet. Velocity fields, shear strain rate and other quantitative flow characteristics can be derived from this particle tracking imaging technique.

PIV investigation of models having curvatures can be difficult due to refraction and reflexion phenomena of the laser beam and also because of the shape itself which makes data aquisition in some areas arduous. These inconvenients result in image distortions and blind spots [5]. Also, manufacturing a flat stent is faster and cheaper than manufacturing a cylindrical stent model for PIV studies. Clearly, when studying various stent design a flat stent configuration could be significantly cost and time efficient when compared to the usual cylindrical geometry.

Similarity technique used to compare and transpose results from a scaled-up flat stent model to an actual cylindrical stent must be done considering the geometric and kinematic differences between them. In this study, the results of a CFD simulation of a flat configuration scaled-up stent model embedded in a rectangular duct to be used in PIV investigation will be compared with the results of a regular cylindical stent geometry. These simulations will take into account the characteristical pulsatilty of the coronary blood flow.

2. Methods

The flat stent model is a scaled-up and unfolded version of implantable endovascular stent [6]. The scaling factor is 12.5, this scaling-up is necessary in order to visualize the flow patterns near the struts using the PIV set-up. The whole strut pattern of the stent is planned to be embedded at the bottom of a transparent polycarbonate rectangular duct. Thus, the width of the rectangular duct corresponds to the scaled perimeter of the cylindrical stent (119.7mm). The height of the rectangular duct is set to be equal to the scaled diameter of the stent (38.1mm) as this offered a similar scaled distance between the stent struts and the maximum velocity located at the middle of the duct for both the rectangle and the cylinder. The following figures show both models and their similar middle plane:

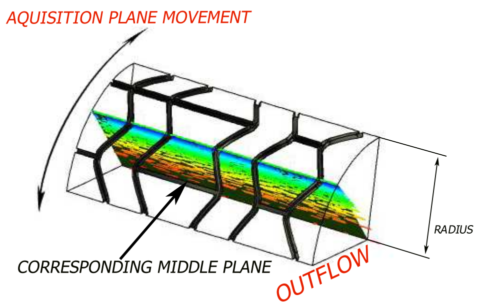

Figure 1: Quarter axi-symmetric cylindrical stent model and its middle plane

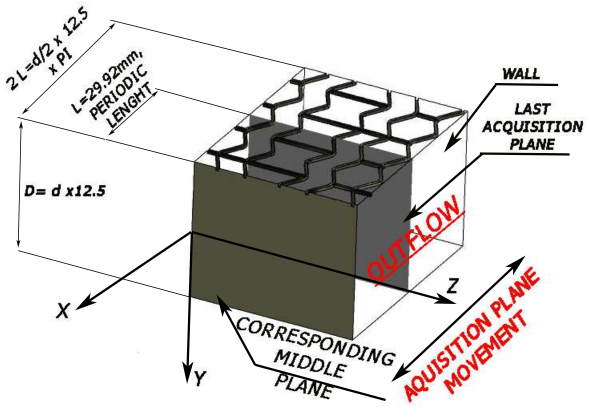

Figure 2: Half flat stent and corresponding middle plane.

The global velocity profile for a fully devellopped flow within a cylinder is given by Poiseuille equation and corresponds to a paraboloid. This profile is different for a rectangular duct. The equations defining this three dimensionnal profile in cartesian coordinates was provided by Lundgren et al. [7].

As shown in Figures 1 and 2 the middle plane of the rectangular duct corresponds to the plane that cuts through the cylindrical CFD model in its middle. In this plane the lateral wall influence on the flow dynamics for the rectangular duct are minimized. Moreover, in this 2-D region the stent geometry of the flat and cylindrical stent are alike. Also, in the rectangular model for the X-Z planes parrallel to the the middle plane the velocity profile given by Lundgren et al. is a cosine series in function of the y axis. Figure 3a) shows that this cosine series function fits the parabolic Poiseuille profile of the cylindrical model with good accuracy (maximum error of 0.05%). Thus, for this similar cutting plane both the stent geometry and the kynematics can be simply scaled from one model to the other with the scaling laws. It then appears natural to take the similar middle planes as a base for the non-dimensionnalization technique and the comparative study between the rectangular and the cylindrical models. In fact, the standard PIV technique is essentially a planar flow investigation technique. It can also be seen from the figure 1 that due to the cyclic strut pattern only a quarter of the stent needs to be investigated. For the flat model this corresponds to a maximum distance of 29.92mm away from the middle duct plane which is the periodic length, L, as shown in figure 2.

The following figure shows the velocity profile of the rectangular duct projected in the Y-Z and X-Z planes respectively.

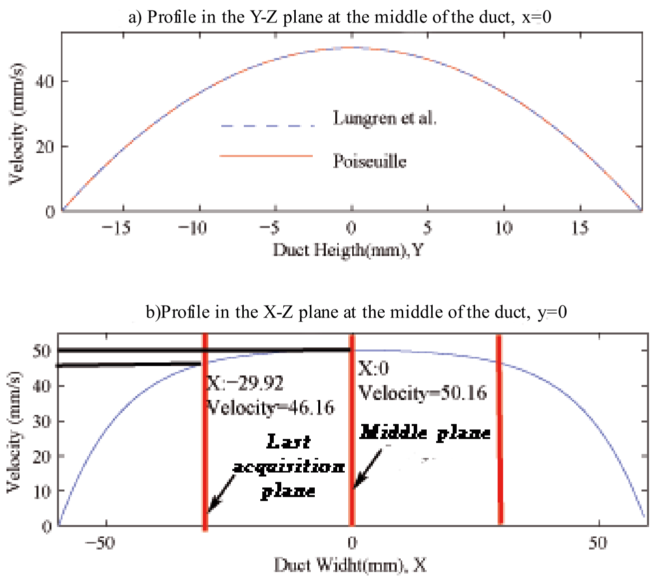

Figure 3: Velocity profile projected on the a) Y-Z plane (Cosines series) and on the b) X-Z plane (hyperbolic cosine series)

Any planes parallel to the initial common plane of the rectangular model will be geometrically similar to their corresponding investigation plane in the cylindrical model. However, the velocity profile will be equivalent to the Poiseuille parabolic 2-D profile of the cylinder with some restrictions. Figure 3 shows that at 29.92mm away from the middle plane, corresponding to the farthest investigation plane, the velocity magnitude is 8% lower than in the middle plane due to side wall effects. The wall effect on the planar velocity field magnitude can easily be quantified by evaluating the hyperbolic cosine series of the velocity profile along the x direction defined by Lundgren et al.. Thus, the underestimation in the results of the shear strain rate at data acquisition planes other than the middle plane can be readjusted accordingly.

The pulsatility of the flow was taken into account by multiplying the 3-D velocity profiles by a standardized velocity waveform [9]. For the flat model this waveform was scaled in time (i.e scaling of the pulse frequency) by maintaining the Womersley number constant for both models.

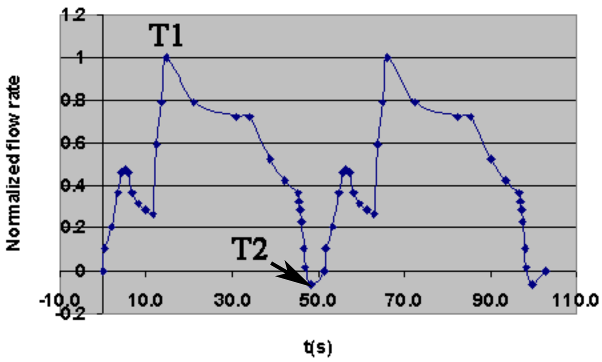

Figure 4: Time scaled standardized flow rate waveform

The most important non-dimensionnal parameters that must be equalized in order to insure similarity between the equivalent data acquisition planes of both models are: the Reynolds number, the Womersley parameter and the loss coefficient. Either the Pi-theorem or the non-dimensionnalization of the Navier Stoke's equations and of Newton's viscous law at vessel wall for a pipe pulsatile flow [8] will yield these adimensionnal numbers. The Reynolds and the Womersley parameters equalized between both models are for average flow condition in the coronary artery [3]. The fluid used for the physiological-cylindrical model is blood. The assumptions for both models include rigid walls, Newtonian, laminar, incompressible and homogeneous flows which have been proved to be sensible when modeling coronary blood flow[9]. The non-dimensionnal numbers of both models appear in the table below.

Parameters	Definition	Cyl.	Rect.
Womersley, α^2	$\frac{\rho\omega D^2}{\mu}$	2.8	2.8
Reynolds, Re	$\frac{\rho D U}{\mu}$	320	320
Loss coefficient, C_f	$\frac{\tau}{\rho U^2}$	-	-

Table 1: Models Parameters

Where:

- ρ, is the density of the fluid
- μ, is the dynamic viscosity of the fluid
- U, is the average velocity in the middle plane
- ω, is the angular velocity or pulsatility of the flow
- D, is the characteristic length being the diameter and the height for the cylindrical and the flat model respectively.
- τ, is the shear stress at the wall

The loss coefficient, C_f, will be used to scale the values of the shear strain rates by the following relation:

$$\dot{\gamma}_{rectangle} = \dot{\gamma}_{cylinder} \frac{\rho_{rectangle} U^2_{rectangle} \mu_{rectangle}}{\rho_{cylinder} U^2_{cylinder} \mu_{cylinder}} \quad (1)$$

The boundary conditions in both CFD models included: a no slip-conditions at the wall, a 3-D velocity profile defined earlier factorized by a pulsatile physiological waveform at the inlet, a zero relative pressure at the outlet and periodicity. The mesh density around the struts and the time step refinement were defined in an earlier CFD simulation of the cylindrical model for the independence of the shear strain rate results throughout the flow domain[6]. This mesh density and time steps were scaled for the rectangular model. The time step used is 0.004 seconds (cylindrical) and the struts height is discretized using 3 prismatic elements for both models.

3. Results

The shear strain rate, the velocity spatial gradient, will be compared at two different instant of the flow: T1, the peak diastolic flow , and T2, the peak systolic flow, of the physiological waveform of figure 4. The wall shear stress between both models will be compared at a line lying at the intersection of the stented wall and the models respective middle plane as shown in Figures 1 and 2. These results appear below.

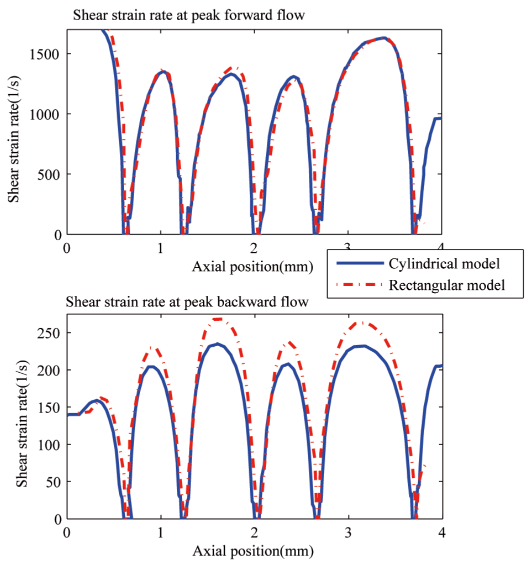

Figure 5: Comparison of shear strain rate values between both models on line 1 at peak diastolic and peak systolic flow.

The following figure shows the variation in time of the shear strain rate for the two geometries compared at a point on the stented wall located at the near the middle of the fourth interstrut position (axial position=3.3mm referring to the cylindrical model appearing on Figure 1.

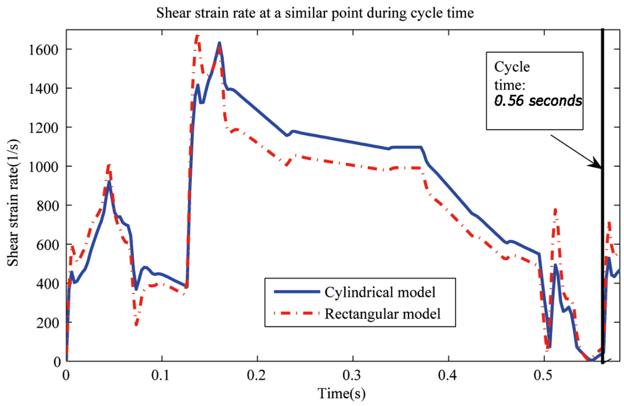

Figure 6: Comparison of shear strain rate values between both models at a similar point throughout the cardiac cycle.

The general map of the shear strain rate for both models at the peak diastolic flow are shown below.

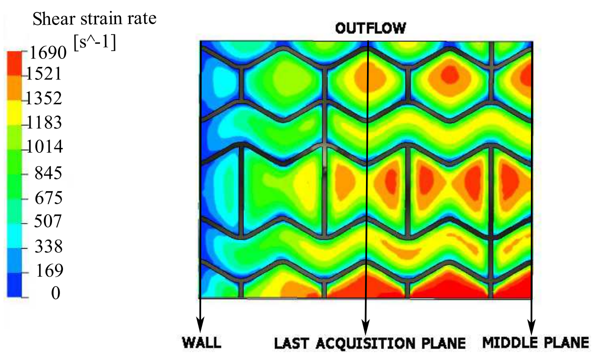

Figure 7: Shear rate at the peak diastolic (forward) flow for the rectangular model.

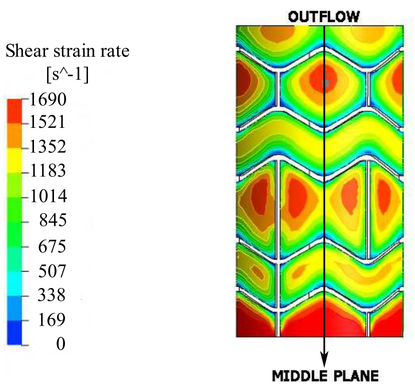

Figure 8: Shear rate at peak systolic (forward) flow for the cylindrical model.

4. Discussion

The results above show a good agreement between the cylindrical and the scaled-up flat model for the shear strain rate derived from the velocity field. Figure 5 shows that for the shear rates along the wall at the common plane the maximum difference between the two simulation is 4% and 14% at peak diastolic and systolic flow respectively. The results of the shear strain rate at a point located on the stented wall through the course of the cardiac cycle appear in Figure 6. It can be seen that the differences between the shear rate at the similar point of interest varies with time throughout the cycle. The table below contains some statistics regarding the variation in the relative difference of the shear rate for the rectangular model with respect to the cylindical model at the point monitored throughout time. The difference in the shear rate values between both model is greater at some few points in time than at peak systolic flow and reach relative differences as high as 150% (0.51s). Despite these few discrepencies between the two simulations the time averaged shear for both model is similar (830 s^{-1} cylinder vs 770s^{-1} rectangle) and the time averaged shear rate relative difference at the point investigated between both model is of 17%.

Time interval (s)	Relative difference (%)
0-0.14	16
0.14-0.28	10
0.28-042.	12
0.42-end	30
Complete cycle	17

Table 2: Relative difference for the shear rate values between both models throughout time for the monitored point

Another interesting fact is that when comparing steady state simulations at flow rates corresponding to a particular instant in the flow the results in shear rate are not the same in many points in time than for the transient case. For example, assuming a steady state flow at the peak systolic flow will yield results in shear strain rate three times lower than in the transient case for the two models studied. Moreover, when comparing the results of shear rates at different flow rates with a steady state flow the relative difference between the rectangular and the cylindrical simulation was below 5%. This confirms the significant influence of the transient behavior of the flow on the wall shear. Figures 7 and 8 illustrate the similarity of the wall shear rate mapping at peak diastolic flow for both models. Regions of low wall shear near the strut edges and higher wall shear in regions further from the struts (middle of stent opened and closed cells) appear in both simulations.

5. Conclusion

A flat model for PIV study has many advantages over a curved stent model. This CFD study confirmed that the physiological shear strain rate values that would occur in a cylindrical stent are retrieved in a rectangular model with a satisfactory relative difference (17% with respect to the cylindical expected values). This was made possible because both models are geometrically and kinematically similar when compared using a planar investigation approach intrinsic to the standard PIV technique. It was also shown in these numerical investigation that the transient behavior of the flow has a significant influence on both the relative difference of the values of shear strain rate between the models and the absolute values of the shear strain rate for a particular model and instant. Further investigations will include PIV studies of both the cylindrical and the rectangular stent models.

References

[1] Serruys PW, de Jaegere P, Kiemeneij F, et al., “A comparison of balloon-expandable-stent implantation with balloon angioplasty in patients with coronary artery disease.” *N Engl J Med*, 331, pp. 489-495, 1994.

[2] Raimund Erbel, M.D., Michael Haude, M.D., Hans W. Hpp, M.D., et al., “Coronary-Artery Stenting Compared with Balloon Angioplasty for Restenosis after Initial Balloon Angioplasty ”*N Engl J Med*, 339, pp. 1672-1678, 1998.

[3] He X, Ku DN.,“Pulsatile flow in the human left coronary artery bifurcation: average conditions.” *J Biomech Eng.*, 118(1), pp. 74-92, 1996.

[4] Benard,N., Coisne,D., Donal,E., Perrault,R., “Experimental study of laminar blood flow through an artery treated by a stent implantation: characterisation of intra-stent wall shear stress” *J.Biomech.*, 36(7), pp. 991-998, 2003.

[5] M.L.Lowe,P.H.Kutt “Refraction through cylindrical tubes” *Experiments in Fluids*, 13, pp. 315-320, 1992.

[6] Faik Issam, “3D Characterization of the Hemodynamics pattern in a Stented Coronary Artery”, *Thesis*, Departement of Mechanical engineering, McGill University, Montreal, Canada, 2005

[7] Barbara Roehrnbauer, “Mathematical and experimental development of two different stent models for PIV application.”, *Thesis*, Departement of Mechanical Engineering, Mcgill University,Montreal,Canada;Central institute for Medical Engineering,TU Munich, Germany, 2005-11-01

[8] F. Muller,W.J.; Gerjarusek,S.; Scherer,P.W., “Studies of wall shear and mass transfer in a large scale model of neonatal high-frequency jet ventilation” *Ann.Biomed.Eng*, 18(1), pp. 69-88, 1990.

[9] Feldman,C.L.; Ilegbusi,O.J.; Hu,Z.; Nesto,R.; Waxman,S.; Stone,P.H., “Determination of in vivo velocity and endothelial shear stress patterns with phasic flow in human coronary arteries: a methodology to predict progression of coronary atherosclerosis” *Am.Heart J.*, 143(6), pp. 931-939, 2002.