Abstract

The stent has been a major breakthrough for the treatment of atherosclerotic vascular disease. The permanent vascular implant of a stent, however, changes the blood flow hemodynamics and may consequently affect the restenosis process. Computational Fluid Dynamics (CFD) has been widely used to analyze the hemodynamic behavior and wall shear stress (WSS) distribution in stented arteries. The objective of this study is to present a thorough comparison among various CFD models to investigate the effects of rheological properties and pulsatile flow on hemodynamic simulation of the intra-stent blood flow. Several CFD models were developed with various modeling setups – axisymmetric parallel ring vs. 3-D stented artery model, Newtonian vs. non-Newtonian flow, and steady-state vs. pulsatile flow. Simulated results show that the minimum WSS occurs at the recirculation zones located at the downstream or backside of each stent struts. The rheological effect on WWS is minor in the axisymmetric parallel ring model; however, it becomes slightly significant in the 3-D stented artery model, with Newtonian flow being a more conservative assumption. For given pulsatile waveforms, the steady-state and pulsatile flow resulted in fairly similar trends in the WSS distribution. Therefore, it is reasonable to simulate the intra-stent blood flow as a steady-state Newtonian flow, which could be beneficial in more complex simulations and drastically reduce the computational time. These findings will provide great insights for future stent design optimization to reduce restenosis.

Keywords: Restenosis; wall shear stress; hemodynamics; coronary artery

1. Introduction

Coronary stenting has become as the primary treatment of cardiovascular diseases and has received great attention from the medical community since its first introduction in 1990s. It is a minimally-invasive method of opening narrowed or blocked coronary arteries that impede blood flow to the heart.
The stent, an artificial wire-mesh tube, provides a scaffolding structure to prevent re-closure of the artery. Over the past two decades, the field has seen numerous innovations in attempts to perfect the percutaneous management of coronary artery diseases. In recent years, stent technology has advanced from bare metal stents to drug eluting stents and biodegradable stents. The development of drug eluting stents resulted in a dramatic reduction of restenosis after angioplasty, leading to worldwide adoption of this technology for coronary artery diseases. Biodegradable stents present the next frontier; they hold the promise of a medical device that could support the artery after intervention, deliver drugs, and disappear without permanently affecting the arteries once its job is done.

Stents are widely used to restore blood flow from atherosclerotic vascular diseases. After stenting, however, patients may develop restenosis, namely the reduction of the artery lumen caused by neo-intimal formation, over a period of several weeks to months. Although restenosis is a complicated process, clinical data have suggested that restenosis is closely related to stent design and local blood flow pattern. Several studies have shown the connection between restenosis rate and stent design [1-2]. Furthermore, a growing consensus suspects that restenosis may be a result of fluid disorder or, more precisely, the variation of wall shear stress (WSS) distribution. Given the role of shear stress in the composition and behavior of atherosclerotic plaques, it is reasonable to assume that it may also be involved in proliferative responses after stenting. Low or oscillating WSS has been observed near locations where neo-intimal thickening in stented arteries is the greatest [3-7]. In addition, several theoretical and in vivo studies have suggested that low or oscillating shear stress, particularly shear stress of less than 5 dyne/cm², may lead to endothelial proliferation of smooth muscle cells [8, 9]. In recent years, numerous Computational Fluid Dynamics (CFD) studies have been conducted with various modeling setups - stationary vs. transient flows, Newtonian vs. complex rheology, or 2-D vs. 3-D. However, only a few papers have made limited comparisons among these different models [10-12].

In this paper, we aim to investigate the effects of rheological properties and pulsatile flow on hemodynamic simulations by presenting a thorough comparison among various modeling techniques. In addition, 3-D simulation was performed to simulate blood flow through an actual stent for comparison. General guidelines for CFD simulations in stented arteries will be concluded to allow higher accuracy and efficiency in future stent hemodynamic analysis.

2. Theory and Method

An axisymmetric CFD model was proposed for hemodynamic analysis, as shown in Fig. 1. The stent is implanted in an idealized straight artery with a uniform diameter $D$, which is assumed to be 3 mm. The stent struts are simplified as a series of semicircles of diameter $d$ with the inter-strut spacing $s$. The artery walls are assumed to be fixed and no-slip. In addition, 3-D simulation was performed to simulate blood flow through an actual stent, as shown in Fig. 2. The diameter of this stent is also 3 mm.

The fluid was assumed to be incompressible, with a density $\rho$ of approximately 1060 kg/m³, and governed by mass conservation and momentum equations:

\[ \nabla \cdot \vec{v} = 0 \]

\[ \rho \left( \frac{\partial \vec{v}}{\partial t} + \vec{v} \cdot \nabla \vec{v} \right) = -\nabla p + \nabla \cdot \tau_D \]
For 2-D simulation, equations were solved by commercial software COMSOL Multi-physics, while another commercial CFD solver, ANSYS FLUENT, was used for 3-D simulation. The typical numbers of nodes in this study were around 9,000 in the axisymmetric model and about 4,700,000 in the 3-D model.

Blood typically behaves like a Newtonian fluid when the shear rate is greater than 100 s$^{-1}$ [13]. However, due to the flow disturbance, the true shear rates in the stented area could be much lower than 100 s$^{-1}$. To understand this non-Newtonian effect, we present a thorough comparison by simulation of Newtonian flow and non-Newtonian flow. For Newtonian simulations, the fluid has a constant viscosity of $\mu = 0.0035$ kg/m-s, while for non-Newtonian simulations, viscosity is described by the Carreau model:

$$\mu = \mu_\infty + (\mu_0 - \mu_\infty)\left[1 + (\dot{\gamma}/\lambda)^n\right]^{\lambda/n - 1}$$  \hspace{1cm} (3)

where $\mu_\infty$ and $\mu_0$ are viscosities as the shear rate goes to infinity and zero, respectively, $\dot{\gamma}$ is the shear rate, and $\lambda$ and $n$ are material constants. These parameters can be inferred by fitting the Carreau model to the experimental data [12-13]. The values obtained from the literature were $\mu_\infty = 0.0035$ kg/m-s, $\mu_0 = 0.25$ kg/m-s, $\lambda = 25.00$ s, and $n = 0.25$. The relationship between the shear rate and viscosity in this Carreau model is shown in Fig. 3.
Oscillatory flow for the pressure gradient of a given frequency in a straight tube was given by Womersley solution [14]. It was assumed that the oscillatory axial pressure gradient in the artery could be expressed as $Ae^{i\omega t}$, where $A$ is a constant, $\omega$ is the angular velocity, and $t$ is the time. Similar to Poiseuille flow, the equation of motion for the uni-directional axial velocity $w$ in a straight tube can be simplified as:

$$\frac{d^2 w}{dt^2} + \frac{1}{r} \frac{dw}{dr} - \frac{\rho \omega^2 w}{\mu} = -\frac{d}{dr}$$

Its solution is written as:

$$w = \frac{A}{r \omega} \left( 1 - \frac{J_0(\alpha r)}{J_0(\alpha R)} \right) e^{i\omega t}$$

where $J_0$ is a zero order Bessel function of the first kind, $r$ is the radial distance, $R$ is the tube radius, $\omega$ is angular frequency, and $\alpha = (\rho \omega / \mu)^{0.5}$ is the Womersley number. For pulsatile flow, the result is obtained by simple superposition of the individual harmonic components of velocity. This solution was used as the inlet condition in the pulsatile simulations in our study. For steady-state simulation the inlet velocity was assumed to be fully developed.

WSS is of significant interest in research dealing with hemodynamics in stented arteries. WSS is defined as:

$$WSS = \frac{1}{r} \left( \frac{\partial u}{\partial r} + \frac{\partial u}{\partial x} \right)_{wall}$$

However, it is much more convenient for quantifying the effects of any characteristic changes if the WSS variation on the arterial wall can be expressed as one single index. In this study, the Average WSS Index (AWI) was used as the major indicator:

$$AWI = \frac{1}{\lambda_{wall}} \int \tau_{wall} dl$$
3. Results and Discussion

Velocity field and the corresponding WSS distribution around the stent area in the axisymmetric model are shown in Figs. 4(a) and (b), respectively. It is apparent that the minimum WSS occurs at the recirculation zones, which are at the downstream or backside of each stent strut.

3.1. Effects of Rheological Properties

Non-Newtonian simulations with the Carreau model were compared with those from Newtonian simulations. Results show that their WSS distributions are nearly the same (Fig. 5(a)). The only difference is the recirculation length reduction in the recirculation zones, by about 3.85% for the case of non-Newtonian flow. The distal recirculation length $L$, a good indicator of neointimal proliferation tendency, is defined as the total distance of negative WSS behind the last strut. At the regions near the backside of each stent strut, the flow velocity and shear rate are fairly low, but a large disparity in viscosity exists between these two models. Despite the viscosity disparity, as illustrated in Fig. 5(b), the wall shear stresses, the products of shear rate and viscosity, are almost identical in these two models in the recirculation zones between struts due to the low shear rate. The viscosity and shear rate simulations are shown in Figs. 5(b) and (c), respectively. In summary, the rheological effects on the wall shear stress are minor in the axisymmetric model.

The effects of non-Newtonian flow become slightly significant but still negligible in the 3-D stented artery model. The WSS contours of the Newtonian and non-Newtonian flow are shown in Figs. 6(a) and (b), respectively. In order to quantify the effects, we measured the total wall surface area with shear stresses lower than 5 dyne/cm², below which restenosis is likely to occur. At the flow rate of 3.11 ml/s, the total wall surface area with shear stresses less than 5 dyne/cm² is 3.83 mm² for the Newtonian flow and 3.55 mm² for the non-Newtonian flow. This result represents a 7.19% reduction in area for the case of non-Newtonian flow. This suggests that the rheological effects on the wall shear stress are still minor, even in the 3-D stented artery model.
Fig. 5. (a) WSS distribution around stent struts (Q = 3.11 ml/s, s/D = d/D = 0.1); (b) Corresponding viscosity distribution around stent struts; (c) Corresponding shear rate distribution around stent struts.

Fig. 6. WSS contours of 3-D simulations for (a) Newtonian flow; (b) Non-Newtonian flow.
3.2. Pulsatile Flow vs. Steady-state Flow

The flow rate in the arteries varies with pulsation. Simulation of this time-dependent pulsatile flow certainly gives a more realistic solution, but also poses a greater challenge. Therefore, in the interests of computational efficiency, it is interesting to compare the pulsatile and steady-state simulations. The velocity profile changes in the right coronary artery (RCA) were obtained from C. Bertolotti et al. (Fig. 7(a)), with the corresponding flow rates at different time instants shown in Fig. 7(b) [14-16]. The key characteristics of this flow are listed in Table 1. To compare the pulsatile and steady-state results, several critical flow rates were chosen for the steady-state simulations (marked as the square dots in Fig. 7(b)).

Figure 8 shows AWI variation in the pulsatile and steady-state simulations. It is apparent that AWI variation basically follows the flow rate changes with only a slight phase difference. In addition, both simulations yield fairly similar trends, with a slightly greater difference near the first peak. Therefore, the steady-state simulation might be a viable alternative model for the pulsatile flow to save computational time with acceptable tolerance.

![Velocity Profiles](image)

![Flow Rate](image)

Fig. 7. (a) Flow rate in coronary artery (one pulse); (b) Velocity profile at different time instants (one pulse).

Table 1. Key features of inlet flow.

<table>
<thead>
<tr>
<th>Feature</th>
<th>Value</th>
</tr>
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<tr>
<td>Time period</td>
<td>0.83 s (72 Hz)</td>
</tr>
<tr>
<td>Max. flow rate</td>
<td>2.20 ml/s</td>
</tr>
<tr>
<td>Min. flow rate</td>
<td>0.31 ml/s</td>
</tr>
<tr>
<td>Mean velocity</td>
<td>16.69 cm/s</td>
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</tbody>
</table>
4. Conclusions

In this paper, we presented a thorough comparison among various CFD models to investigate the effects of rheological properties and pulsatile flow on hemodynamic simulations. For both the axisymmetric model and 3-D stented artery model, since the rheological properties do not produce significant effects, it is reasonable to simulate the blood flow with the assumption of Newtonian flow. Furthermore, it is also reasonable to assume the blood flow as steady-state in the axisymmetric model, as both pulsatile and steady-state simulations basically yield similar trends. This assumption could help drastically reduce the computational cost. It is hoped that the discoveries in this paper could provide great insights for future stent design optimization to reduce restenosis.

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References


