STENT GEOMETRY AND DEPLOYMENT RATIO INFLUENCE DISTRIBUTIONS OF WALL SHEAR STRESS: THREE-DIMENSIONAL NUMERICAL SIMULATIONS EXPLOITING PROPERTIES OF AN IMPLANTED STENT

John F. LaDisa, Jr. (1,2), David C. Warltier (1,2), Ismail Guler (3), Judy R. Kersten (2), and Paul S. Pagel (1,2)

(1) Department of Biomedical Engineering, Marquette University, Milwaukee, WI

(2) Department of Anesthesiology, Medical College of Wisconsin, Milwaukee, WI

(3) Boston Scientific - Scimed, Maple Grove, MN

INTRODUCTION

An estimated 2 million stents are deployed annually to restore blood flow distal to vascular stenoses[1]. Unfortunately, in-stent restenosis occurs in approximately 20% of patients. Rates of restenosis after stent implantation vary with stent type. Stent geometry has been identified as an important predictor of neointimal hyperplasia[2]. Vascular geometry is known to profoundly influence local shear stress distributions as branching and curvature adversely affect the ideal flow environment and correlate with sites of intimal thickening. Interestingly, stents have also been shown to influence coronary arterial hemodynamics and wall shear stress during conditions of resting and elevated blood flow *in vivo* and using computational fluid dynamics (CFD) modeling[3,4]. Excessive expansion of a stent during deployment may also cause extensive vessel damage, and recent data indicate that shear stress mediates the rate of smooth muscle cell migration after injury[5]. Several models *in vivo* have adopted a stent to artery diameter ratio of 1.1-1.2 to 1 to limit such vascular damage during deployment. Nevertheless, the theoretical influence of stent design (including the number, width, and thickness of struts comprising the stent) and the influence of excessive stent expansion on distributions of wall shear stress have not been comprehensively evaluated. We tested the hypothesis that geometric properties and extent of expansion of a deployed stent differentially influence distributions of wall shear stress predicted using three-dimensional CFD modeling.

METHODS

Construction of Stented Vessels

CFD models were created from measurements of canine left anterior descending (LAD) coronary artery blood flow and diameter using a transit-time flow probe and ultrasonic segment length transducers, respectively[4]. After a baseline equilibration period, blood flow and diameter waveforms were digitized (364 Hz) in the presence and absence of a 16 mm slotted tube stent. Stented vessels were simulated using a geometric construction and mesh generation algorithm custom designed using Matlab. Computational vessels were composed of structured hexahedral control volumes arranged in a four-domain butterfly design that exploited symmetric stent and vessel properties to model half of the computational vessel. Computational vessels were created consisting of 2, 4 or 8 struts using two strut widths (0.15 or 0.35 mm) and two different degrees of protrusion into the flow domain (0.1 or 0.075 mm) (Table 1). Computational geometries used to examine the impact of over-expanded stents were created using stent-to-vessel ratios of 1.1 and 1.2 to 1. All computational stents were 16 mm long and the diameter of the native vessel for all simulations was 2.74 mm.

Table 1. Stent and Vessel Properties

Simulation Number	Number of Struts	Strut Width (mm)	Strut Thickness (mm)	Stent-to-Artery Ratio
1	2	0.35	0.1	1 - 1
2	4	0.35	0.1	1 - 1
3	4	0.15	0.1	1 - 1
4	4	0.35	0.075	1 - 1
5	8	0.15	0.1	1 - 1
6	8	0.15	0.075	1 - 1
7	4	0.35	0.1	1.1 - 1
8	4	0.35	0.1	1.2 - 1

Computational Model Simulations

Simulations were performed using the commercially available software package CFD-ACE (CRDRC; Huntsville, AL). This software uses a finite volume approach to solve the Navier-Stokes equations at the center of each hexahedral control volume. Maximum blood flow velocity during diastole (0.225 m/s) was imposed as a parabolic velocity profile at the inlet of each geometry. Computational simulations were conducted assuming a Newtonian, incompressible fluid with a density of 1060 kg/m^3 and viscosity of 3.7 cP.

Quantification of Simulation Results

Wall shear stress was determined as the product of viscosity and shear rate. Shear rate was calculated from the second invariant of the strain rate tensor. Vascular pathology associated with neointimal hyperplasia correlates with geometrically specific deviations in wall shear stress from a normal range as regions of high shear stress protect against hyperplasia while areas of low wall shear stress strongly

correlate with regions of intimal thickening. Thus, a wall shear stress value associated with vascular remodeling (15 dynes/cm^2) was used as the threshold for comparison between simulations[6]. Images of simulation results were spatially aligned and the distribution of shear stress divided into 12 grayscale levels in 3 dynes/cm^2 intervals. Digital image processing was performed to threshold wall shear stress values ≤15 dynes/cm^2 and the area quantified using ImageJ (NIH; Bethesda, MD). This area was then expressed as a percent of the stent and total vessel area.

RESULTS

Figure 1 illustrates distributions of wall shear stress in a typical CFD simulation. The highest values of wall shear stress were localized over the surface of stent struts. In contrast, regions of low wall shear stress surrounded the stent struts independent of the strut width, thickness, number, or stent-to-vessel diameter ratio. The ratios for the area of wall shear stress ≤15 dynes/cm^2 with respect to the stent and total vessel area for each simulation are summarized in table 2. The area of computational vessels subjected to wall shear stress ≤15 dynes/cm^2 increased as the number of struts increased. Increases in stent width and thickness also caused the distribution of wall shear stress in this range to increase.

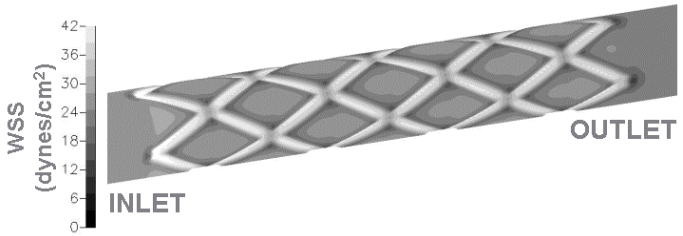

Figure 1. Distributions of wall shear stress (WSS) within the stented segment of a computational vessel.

Table 2. Wall Shear Stress, Stent, and Vessel Area Analysis

Simulation Number	Total Area (mm^2)	Stent Area (mm^2)	WSS Area ≤15 dynes/cm^2 (mm^2)	Stent/ Total Area (%)	WSS/ Stent Area (%)	WSS/ Total Area (%)
1	62.2	6.35	1.15	10	18	2
2	60.8	11.4	3.27	19	29	5
3	60.8	9.13	0.71	15	8	1
4	60.8	10.9	1.12	18	10	2
5	62.2	15.1	5.78	24	38	9
6	62.2	14.1	2.21	23	16	4
7	66.5	11.4	31.0	17	273	47
8	73.0	11.4	46.8	16	411	64

Simulations with four struts revealed that reducing the strut width by more than half reduced the area of wall shear stress ≤15 dynes/cm^2 to a greater extent then a 25% reduction in strut thickness. Lower stent-to-vessel ratios correlated ($R^2 = 0.54$) with a reduction in areas of shear stress below the threshold value. Over-expansion of the stent had a profound impact on wall shear stress distributions and vastly increased the area of wall shear stress ≤15 dynes/cm^2 compared to stents implanted in a 1 to 1 stent-to-vessel ratio. Over 55 and 87% of the local stented segment was subjected to distributions of wall shear stress in this range for the simulations with deployment ratios of 1.1 and 1.2 to 1, respectively.

DISCUSSION

In-stent restenosis remains an important problem despite the ability of stents to restore vascular blood flow in the majority of cases. Collective evidence suggests that rate of restenosis varies with stent type, that local vessel geometry correlates with regions of intimal hyperplasia, and that vascular response to injury is modulated by shear stress. These data imply that the geometry of the implanted stent may cause altered distributions of wall shear stress that influence the development of neointimal hyperplasia. The present results indicate that the number, width, and thickness of stent struts cause distinctive patterns of wall shear stress. These results are consistent with previous studies using flow visualization through Wallstents of various mesh densities and a CFD analysis of 2-D stent strut spacing[7]. The results further indicate the reducing the stent to vessel area may independently reduce areas of disturbed wall shear stress patterns as was suggested by a previous study examining intimal hyperplasia in rabbit iliac arteries[8]. Recently Liu *et. al.*[5] demonstrated that the rate of smooth muscle cell migration was greatest in regions of lower shear stress after vascular injury. Thus, vascular damage produced by an over-expanded stent may predispose this vascular segment to the development of neointimal hyperplasia by adversely affecting distributions of wall shear stress.

In summary, the present results using three-dimensional CFD modeling predict that geometric properties of an implanted stent cause unique distributions of wall shear stress. These theoretical results further suggest that stent geometries that minimize stent to vessel area, and reduce strut number, thickness, and width appear to be less likely to expose the vessel to distributions of shear stress that have been implicated in neointimal hyperplasia.

REFERENCES

1. Holmes, D., HJ, FD, VR, JA, and KS, "ACC expert consensus document of coronary artery stents," Journal of the American College of Cardiologists, vol. 32, pp. 114-182, 1998.
2. Kastrati, A., Mehilli, J., Dirschinger, J., Pache, J., Ulm, K., Schuhlen, H., Seyfarth, M., Schmitt, C., Blasini, R., Neumann, F. J., and Schomig, A., "Restenosis after coronary placement of various stent types," American Journal of Cardiology, vol. 87, pp. 34-39, 2001.
3. LaDisa, J. F., Olson, L. E., Hettrick, D. A., Guler, I., DiCaprio, F., Pagel, P. S., Kersten, J. R., and Warltier, D. C., "Computational fluid dynamics modeling of hemodynamic changes caused by coronary stenting in vivo," Annals of Biomedical Engineering, vol. 29, pp. S28, 2001.
4 LaDisa, J. F., Hettrick, D. A., Olson, L. E., Guler, I., Gross, E. R., Kress, T. T., Kersten, J. R., Warltier, D. C., and Pagel, P. S., "Coronary stent implantation alters coronary artery hemodynamics and wall shear stress during maximal vasodilation," Journal of Applied Physiology, vol. 93, pp. 1939-1946, 2002.
5. Liu, S. Q. and Goldman, J., "Role of blood shear stress in the regulation of vascular smooth muscle cell migration," IEEE Transactions on Biomedical Engineering, vol. 48, pp. 474-483, 2001.
6. Malek, A. M., Alper, S. L., and Izumo, S., "Hemodynamic shear stress and its role in atherosclerosis," JAMA, vol. 282, pp. 2035-2042, 1999.
7. Berry, J. L., Santamarina, A., Moore, J. E., Roychowdhury, S., and Routh, W. D., "Experimental and computational flow evaluation of coronary stents," Annals of Biomedical Engineering, vol. 28, pp. 386-398, 2000.
8. Garasic, J. M., Edelman, E. R., Squire, J. C., Seifert, P., Williams, M. S., and Rogers, C., "Stent and artery geometry determine intimal thickening independent of arterial injury," Circulation, vol. 101, pp. 812-818, 2000.