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Combined effects of pulsatile flow and dynamic curvature on wall shear stress in a coronary artery bifurcation model

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Abstract

A three-dimensional model with simplified geometry for the branched coronary artery is presented. The bifurcation is defined by an analytical intersection of two cylindrical tubes lying on a sphere that represents an idealized heart surface. The model takes into account the repetitive variation of curvature and motion to which the vessel is subject during each cardiac cycle, and also includes the phase difference between arterial motion and blood flowrate, which may be nonzero for patients with pathologies such as aortic regurgitation. An arbitrary Lagrangian Eulerian (ALE) formulation of the unsteady, incompressible, three-dimensional Navier–Stokes equations is employed to solve for the flow field, and numerical simulations are performed using the spectral/hp element method. The results indicate that the combined effect of pulsatile inflow and dynamic geometry depends strongly on the aforementioned phase difference. Specifically, the main findings of this work show that the time-variation of flowrate ratio between the two branches is minimal (less than 5%) for the simulation with phase difference angle equal to 90°, and maximal (51%) for 270°. In two flow pulsatile simulation cases for fixed geometry and dynamic geometry with phase angle 270°, there is a local minimum of the normalized wall shear rate amplitude in the vicinity of the bifurcation, while in other simulations a local maximum is observed.
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1. Introduction

It is widely believed that the development and progression of atherosclerosis is related to the complex flow field occurring in curvatures and bifurcations of large and medium arteries (Caro et al., 1971; Ku et al., 1985; Friedman et al., 1987). Numerical studies based on rigid, idealized (Steinman et al., 2000; Bertolotti and Deplano, 2000) as well as realistic geometries (Myers et al., 2001; Steinman et al., 2002; Zeng et al., 2003) have been useful in understanding the unique complex features of these flows. Most of these investigations focus on the effects of the geometry, pulsatile flow and non-Newtonian behavior of blood in non-moving vessels. Perktold and Rappitsch (1995) and Zhao et al. (2000) analyzed the effect of distensible wall on the local flow field. Some authors have suggested that the flexibility and motion of coronary artery during each contraction and expansion of the heart is important in the study of the flow dynamics. Several investigations used models in which the vessel was represented by a single flexible tube (Schilt et al., 1996; Santamarina et al., 1998; Zeng et al., 2003).

Comparatively less is known about the effects of vessel movement on the blood flow patterns in coronary arteries at *bifurcations*. Zeng et al. (2003) simulated a realistic arterial motion based on biplane cineangiograms

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in human right coronary arteries but did not consider any branches. Weydahl and Moore (2001) were among the first to consider the model of coronary artery at bifurcation with time-varying geometry. They showed that curvature variation is important in determining temporal wall shear stress variations. However, they considered only steady inflow.

The objective of the present work is to analyze the *combined effect* of dynamic geometry and pulsatile inflow on the flow dynamics and wall shear rate in a simple model of a coronary artery at bifurcation. This will allow us to take into account any *phase difference* between the two unsteady phenomena that may arise in cardiac disease.

Normal coronary flow velocities are characterized by a small forward flow during systole and a large forward flow during diastole (Guyton, 1986). In cardiac disease other patterns can arise. For example, if there is an infarcted region of myocardium, the cardiac muscle in that region does not contract as healthy muscle. Consequently, the epicardial blood flow variation and the deformations of the ventricular wall masses may both change. If the infarct is due to current thrombotic obstruction, outflow resistance in the corresponding intramural segments is not relieved in diastole. Matsuo et al. (1988) have connected the change in phasing information to aortic regurgitation. Using a bidirectional Doppler flowmeter catheter to examine coronary flow velocity in patients who had aortic valve disease, they found a decreased diastolic and increased systolic coronary flow. In some cases, the diastolic flow velocity was less than the systolic. These flow velocity patterns became notable with severe aortic regurgitation. Such phenomena have a direct effect on the *phase difference* between the arterial motion studied here and the pulsatile inflow imposed in our models.

While approximate values for epicardial artery curvature and its variation through a heartbeat are known, precise values are problematic to obtain. Biplane or multiplane angiograms taken in vivo are useful for estimating the axes of vessel segments through the cardiac cycle but do not register lumenal diameters or details sufficiently accurately, and do not show artery wall tissue characteristics. Intravascular ultrasound can supply lumenal geometry and wall tissue information but does not of itself find vessel curvature, takes some time to generate in vivo, and is limited to larger-diameter coronary vessels. MRI, while potentially providing geometric information and wall structure details for coronary arteries, is still not at a state-of-art where spatial resolution is as good as desired for the present study. Thus, to explore fluid dynamic features of curvature variation as well as curvature and flow pulsatility we have chosen to examine flow in a branched vessel model with prescribed but representative parameter values.

2. Basic assumptions

A three-dimensional bifurcating model is defined by an analytical intersection of two cylindrical tubes lying on a sphere that represents an idealized heart surface (Fig. 1). Due to the sharp edge at the junction of arterial branches the solution has a numerical singularity that is localized in a small region close to the junction and does not seem to affect the results in other parts of the domain. The heart motion is simulated by changing the sphere radius, R; the center of the sphere is fixed at the coordinate origin. Gross and Friedman (1998) obtained the dynamics of coronary artery curvature from biplane cineangiograms. The results of this study suggest significant harmonic content up to 6 Hz in curvature variation, and this was taken into account in the numerical studies of Weydahl and Moore (2001) and Moore et al. (2001). Here we have adopted the same range of parameters as in Weydahl and Moore (2001) and Moore et al. (2001), i.e. we take the frequency of the sphere radius variation to be 5 Hz, so that the period $T = 0.2$ s. More specifically, R is specified as a sinusoidal function

$$R(t) = R_0(1 + \delta \sin(2\pi t/T)),$$

where the mean sphere radius R_0 is set to 56.25 mm. Three different values of the parameter δ were used in simulations, 0.0, 0.1 and 0.3. In addition, two cases were considered with $\delta = 0.0$ and R equal to 50.625 and 61.875 mm, i.e., minimum and maximum radii for the dynamic case with $\delta = 0.1$.

The tubes have a circular cross section with constant diameters $D_1 = 3$ mm and $D_2 = 1.5$ mm. At time $t = 0$, the length of the segments AB, BC and BD are equal to 10.125, 24.0 and 12.375 mm, respectively. The lengths of arterial segments are fixed in time so that the total volume of the model remains constant. The junction angle θ is 45°, which is different from the bifurcation modeled in Weydahl and Moore (2001). The axis of the large tube is located in the xy-coordinate plane while the small tube is in the $z > 0$ half-space. The point of intersection of the tube axes B lies on the x-coordinate axis during the entire cardiac cycle. The blood is assumed to be an incompressible, Newtonian

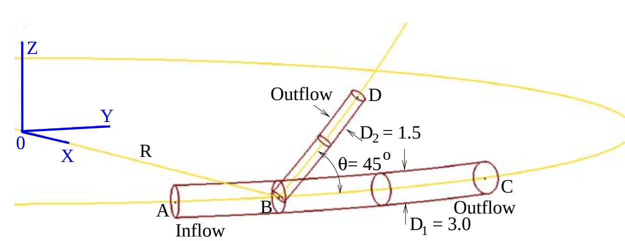

Fig. 1. Geometry of the bifurcation and coordinate system (dimensions in mm).

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