The Combined Effect of Pulsatile Inflow and Unsteady Geometry on Flow in Coronary Arteries

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Introduction

Numerous investigations suggest that the complex flow fields occurring in medium and large arteries are related to the development and progression of atherosclerosis. Coronary arteries are among the most commonly diseased vessels. There are indications that flow pulsatility, as well as flexibility and motion of coronary artery during each contraction and expansion of the heart are important in the study of the flow features. The intention of the present work is to analyze the combined effect of pulsatile inflow and unsteady geometry on flow in a model of coronary artery at bifurcation.

Methods

A three-dimensional model of a coronary artery at bifurcation is built as an analytical intersection of two cylindrical tubes which lie on a sphere that represents an idealized heart surface. The tubes have a circular crossection with fixed diameters ($D_1 = 3\ mm$ and $D_2 = 1.5\ mm$) and their lengths are constant in time. The junction angle is equal to 45^0. The heart motion is simulated by changing the sphere radius, R.

In [1] the dynamics of coronary artery curvature was obtained from biplane cineangiograms. The results suggested that there is a significant harmonic content up to 6 Hz in curvature variation [2]. Following [2], we assume the frequency of sphere radius variation to be 5Hz. More specifically, R is specified as $R(t) = R_0(1 + \delta \sin(10\pi t))$. The mean sphere radius R_0 is set to $56.25\ mm$. Instead of rather complicated physiological waveform of coronary flow velocity, we use a simple time-dependent sinusoidal function. At the *inflow* the pulsatile flat velocity profile is specified, $U(t) = U_0(1+\epsilon \sin(10\pi t + \frac{\alpha\pi}{180}))$, $U_0 = 400\ mm/s$. Four different values of the phase difference angle α are used in simulations. At the two *outflow* sections (main and side branch) a constant pressure and zero normal derivatives of velocity are imposed. No-slip conditions are used at the vessel walls.

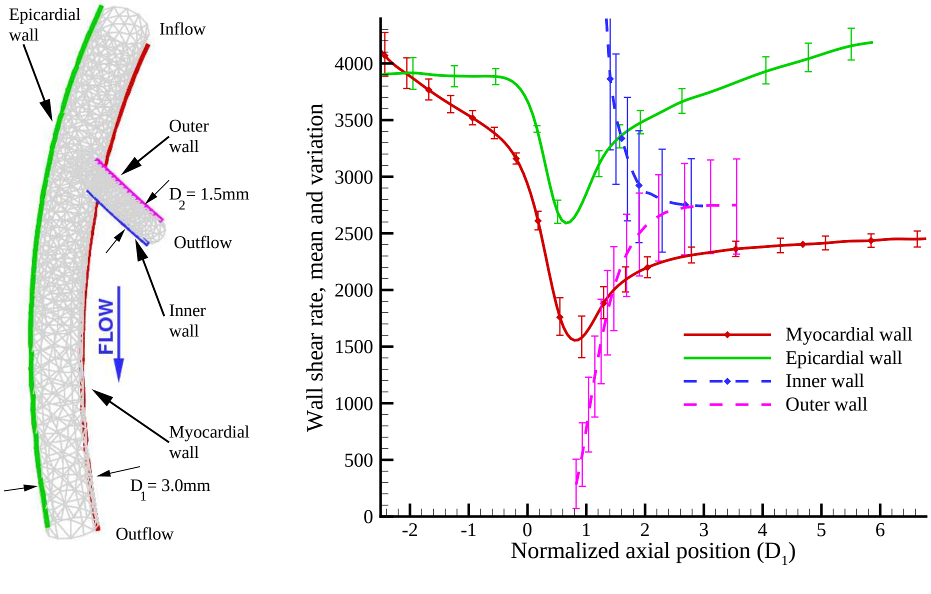

Figure 1: Wall shear rate, mean and variation, extracted along selected lines on different walls of the dynamically deformed model with steady inflow($\delta = 0.1, \epsilon = 0.0$). The distance is measured from the intersection point of the tube axeses and normalized by the large tube diameter D_1.

The blood is assumed to be an incompressible, Newtonian and homogeneous fluid. The flow is three-dimensional and unsteady. The Reynolds number, based on the main branch diameter D_1 and mean inflow velocity U_0, is equal to 300, which is a typical value for blood flow in large arteries. An arbitrary Lagrangian Eulerian (ALE) formulation of unsteady, incompressible, three-dimensional Navier-Stokes equation is employed to solve the flow field, while a velocity smoothing method is used for updating the computational mesh. Numerical simulations are performed using the spectral/hp element solver Nektar [3].

Results and discussion

The effect of dynamic geometry($\delta = 0.1, \epsilon = 0.0$) on the wall shear rate (WSR) is shown on Figure 1. The mean (time average) and variation of WSR during one periodic cycle are extracted along selected lines on the myocardial and epicardial walls of the main branch, and the inner and outer walls of the side branch. The variations of WSR along the myocardial wall are significant in comparison to its low mean. The low values of mean WSR are also observed on the outer wall below the bifurcation. The region of the largest variation of WSR during the cycle is located on the surface of the side branch. This can be attributed to the variation (up to 26%) of the flowrate through the branch, which is likely due to the motion of the artery as it lies on the surface of the simulated heart. The difference in flowrate through the side branch is found to be insignificant in quasi-static simulations, when the geometry is fixed with mean, minimum and maximum curvature radii.

To characterize the variation of WSR during the cycle we use the normalized wall shear rate amplitude(NWSRA). NWSRA was introduced in [4] by Santamarina et al. It is defined as a difference between the maximum and minimum values of the WSR during the cycle, devided by the mean($\delta = 0.0, \epsilon = 0.0$) WSR. Figure 2 demonstrates the effect of curvature, inflow velocity and phase difference angle α variations on NWSRA. The latter, α, has a remarkable influence.

In conclusion, it is evident from these studies that the combined effect of pulsatile inflow and dynamic geometry has significant influence on the flow dynamics and wall shear rate in a model of a coronary artery at bifurcation.

Acknowledgements

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References

[1] M.F. Gross, M.H. Friedman. J. of Biomechanics,1998;31:497.
[2] E.S. Weydahl, J.E. Moore. J. of Biomechanics,2001;34:1189.
[3] G.E. Karniadakis, S.J. Sherwin. Spectral/hp Element Methods for CFM. Oxford University Press, 1999.
[4] A. Santamarina, E. Weydahl, J.M. Siegel, J.E. Moore. Ann. of Biomedical Engineering,1998;26:944.

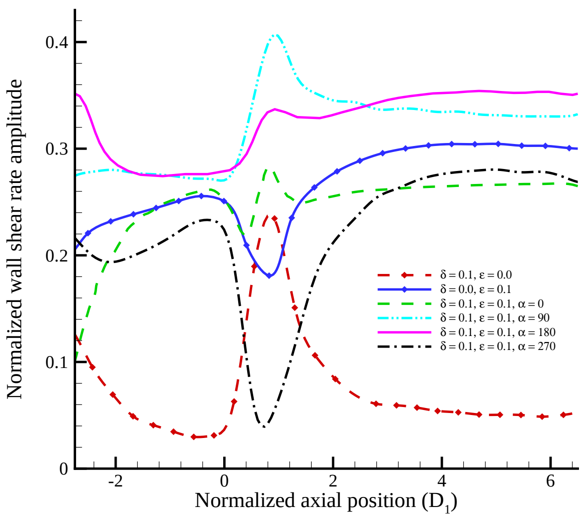

Figure 2: Normalized wall shear rate amplitude extracted along the myocardial wall for different simulation cases.