Numerical study on effects of Newtonian and Non-newtonian blood flow on local hemodynamics in a multi-layer carotid artery Bifurcation

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Abstract

Many cardiovascular diseases, such as atherosclerosis, frequently occur on the Carotid ar tery bifurcation. Hemodynamic factors are crucial in the functioning of the cardiovascular system; therefore, Finite element simulations of fluid–solid interactions were used to investigate variations in flow dynamics as well as wall mechanics at a layered carotid artery bifurcation, and its effects on atherogenesis in Ideal Carotid bifurcation. In the present study, a rigid-wall model and a two FSI models are studied to find out how much the mechanics of wall affects the hemodynamics' distribution. In the computations, blood was modeled as both Newtonian and non-Newtonian fluid. The shear thinning behaviour of blood was incorporated through the Carreau Yasuda model. Under the steady boundary conditions which are applied here, though deformability changes the hemodynamic factors significantly, different rheological model plays a more significant role compared to deformability.

Keywords: Computational Fluid Dynamics, Layered carotid artery bifurcation , Fluid-Structure Interaction, Newtonian- non Newtonian, Computational Cyclic strain

1. Introduction

Arterial diseases, namely atherosclerosis, are the leading cause of death in modern societies [1]. One of the most prone sites of atherosclerosis in arterial tree is carotid artery bifurcation [2]. It provides brain and other facial organs with blood hence it is a very critical zone. It is believed that local hemodynamic is in close relation with initiation and progression of this disease. To date many studies have focused solely on the role of rheological models that describe the transport properties of blood on distribution of hemodynamic factors [3, 4].

These studies have been performed in order to explore the role of hemodynamics in the formation of stenosis. But much less studies have considered the effects of deformability of arterial wall [5]. Flow field characteristic is the result of the fluid interaction with the solid boundary of the vessel wall. In addition, many hemodynamic factors such as circumferential strain and stress phase angle that are said to be highly contributed to plaque formation [5], cannot be defined and studied unless deformability of vessel wall is added to the model.

This study deals with the steady state of the blood flow in a three-dimensional and layered model of the carotid artery bifurcation. In this part, blood is modelled as both Newtonian and non-Newtonian fluid. Carreau-Yasuda model is used to describe the transport properties of blood. A comparison between numerical results of Newtonian and non-Newtonian hemodynamic is presented.

2. Materials and method

2.1 Governing equations

The numerical modelling of incompressible and isothermal flow of fluid requires the solution of Navier-Stokes equations and the continuity equation as well. The governing equations are the three-dimensional incompressible Navier–Stokes equations and are given as:

$$\nabla.\overrightarrow{v_f} = 0 \tag{1}$$

$$\rho\left[\frac{\partial\overrightarrow{v_f}}{\partial t} + ((\overrightarrow{v_f} - \overrightarrow{v_s}).\nabla)\overrightarrow{v_f}\right] = -\nabla p + \nabla.\vec{\tau} \tag{2}$$

Where v, ρ represent the velocity and density respectively. For FSI modelling, these equations have to be solved in company with governing equations of solid domain which is:

$$\frac{\partial\rho_s}{\partial t} + \nabla.\rho_s\overrightarrow{d_s} = 0 \tag{3}$$

$$\rho_s\vec{a}_s = \nabla.\vec{\sigma}_s + \vec{f}_s \tag{4}$$

In order to be able to evaluate and verify the results with previous studies [4], blood is modelled with the exact same physical and transport properties in the mentioned reference.

Equation 5 represents the mathematical models describing the non-Newtonian transport properties of the blood.

$$\frac{\mu-\mu_\infty}{\mu_0-\mu_\infty} = [1 + (\lambda\dot{\gamma})^a]^{\frac{(n-1)}{a}} \tag{5}$$

$\mu_0 = 22 \times 10^{-3}\ Pas, \mu_0 = 2.2 \times 10^{-3}\ Pas, a = 0.644,$
$n = 0.392, \lambda = 0.110\ sec$

2.2 Boundary conditions

In addition to the boundary conditions used for fluid domain, Boundary Conditions at interface of solid and fluid needs to specified. We specified that not only displacement and velocity of both solid and fluid domains are needed to be the same at interface, but also the fluid and solid domain share the same value of wall shear stress. At inlet of CCA a fully developed parabolic velocity profile is applied while both outlets are treated as traction free boundaries. The lengths of two branches were manipulated to reach the desired flow division ratio which is 0.45[4].

2.3 Geometry

The geometry used in this study is the same geometry used by [4]. In this way the results of present study could be easily compared and verified. The general shape of ideal Carotid bifurcation is shown in fig.1. Besides two sections and different parts of this geometry can be seen in fig 1:

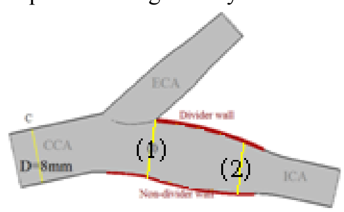

Fig. 1. Geometry of the ideal carotid artery bifurcation and velocity measurement zones

Vessel wall is modelled as an isotropic elastic material with uniform thickness of 0.7 mm. The stress-strain behaviour of vessel wall can be seen in fig2. According to previous studies [6], the maximum strain of carotid artery bifurcation falls before shear stiffening area. So assumption of linear elastic seems to be a good and exact assumption for arterial wall mechanics. Because the elastic modulus in circumferential and axial directions are pretty close (table 1), we can claim that an isotropic model can describe solid domain very well.

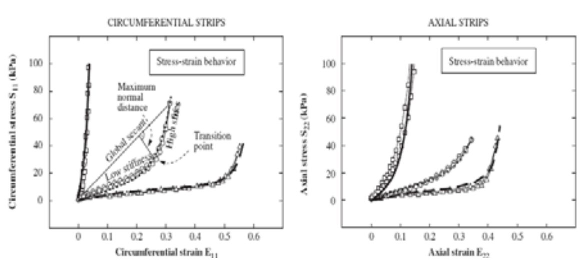

Fig. 2. Stress strain behaviour of vessel wall [7]

Table 1- Mechanical Properties of vessel wall in circumferential(Left) and axial(Right) directions[7]

Intima	112 kPa	80 kPa
Media	22 kPa	15.8 kPa

3. Results and conclusion

The validity of the study was checked and velocity profiles at different measurement zones were compared with previous experimental and numerical studies.

In the following figures, the vector plot of non Newtonian and Newtonian models are presented for layered model. As can be seen in both layered and rigid models, Newtonian model undergoes higher levels of complexity and curvature as well as larger separation and back flow zones. This behaviour could be easily predicted, because that viscosity in Newtonian model experiences lower values compared with the apparent viscosity in Carreau-Yasuda model

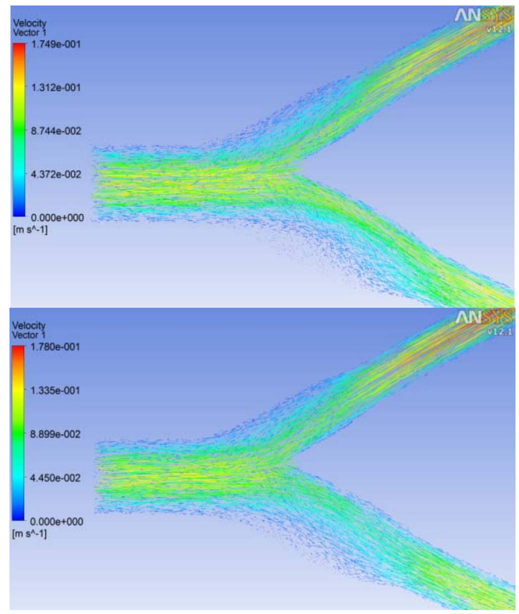

Fig. 3. Velocity vector plot for Newtonian(Above) and Non Newtonian (Below) for flow in layered model

as a non-Newtonian model, especially at the lower shear rates. Implementing Newtonian model results in an underestimation in apparent viscosity that leads to an overestimation in the size of back flow zone.

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