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Blood Analog Fluid Flow in Vessels with Stenosis: Development of an Openfoam Code to Simulate Pulsatile Flow and Elasticity of the Fluid

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Abstract

The present work reports a numerical study of the flow of a blood analog fluid in vessels with a stenosis using OpenFOAM code. The CFD package OpenFOAM was selected to perform the numerical study because it is an open software package that can be freely modified. Pulsatile flow and elasticity of a blood analog fluid were considered and implemented in a numerical code. Velocity profiles were obtained for an artery with a stenosis, considering constant inlet velocity and elasticity of blood analog fluid. The profiles obtained through OpenFOAM are in agreement with those obtained through the analytical solution. Velocity profiles were also obtained for a bifurcation with a stenosis, considering pulsatile flow and shear-thinning of the fluid. The profiles obtained through OpenFOAM are in agreement with those obtained by ANSYS.

Keywords: Pulsatile flow, viscoelasticity, stenosis, bifurcation, OpenFOAM

1. Introduction

Cardiovascular diseases, caused by stenoses or aneurysms, are one of the main causes of death worldwide [1]. Therefore, the study of human blood flow has gained great importance and can contribute, as an auxiliary tool, to the prevention and treatment of cardiovascular diseases. Several authors have been studying,

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numerically, perturbations in human blood flow: Botar et al. [2] studied the portal vein blood flow, Scotti et al. [3] studied blood flow in an aortic aneurysms, Banerjee et al. [4] studied blood flow in stenotic arteries and Sousa and Castro et al. [5, 6] the flow in the carotid bifurcation.

However, as far as we know, the study of blood flow in a bifurcation with a stenosis has not yet been explored in the literature. Stenoses frequently occur after the main bifurcation of the left coronary artery. These stenoses can be responsible for the obstruction of the blood flow, and so can be the cause of death.

Authors of the present paper started to study the flow of a blood analog in a bifurcation with a stenosis (designed in CAD system) through numerical methods using ANSYS [7]. In that study, the shear-thinning of blood was considered but the elasticity of the fluid was neglected. Moreover, steady state was considered and pulsatile flow was neglected.

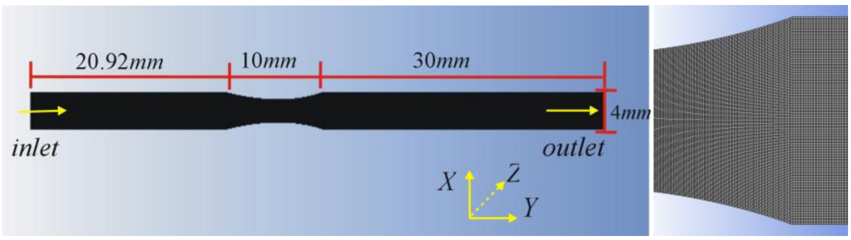
The present work is an evolution of the work presented by Pinto et. al. [7]. The final objective of the authors is to simulate blood flow, considering, simultaneously, complex geometries (bifurcation case), pulsatile flow and both elasticity and shear-thinning behaviors. The implementation of the fluid elasticity in a CFD package requires the direct access to the source code controlling the specification of the equations and the numerical methods. For this reason the open source package OpenFOAM was used. However, the convergence taking into account all the considerations, simultaneously, is very difficult. The problem of convergence must be solved step by step. In the present work, results are obtained, firstly, for an artery with a stenosis, and, then, for a bifurcation with a stenosis.

2. Procedure

2.1. Computational Mesh

Figure 1a represents the geometry, dimensions and computational grid of an artery with a stenosis – XY plane slice. The mesh used is very refined (Figure 1b) in order to obtain convergence. For the artery with a stenosis, the following conditions were taking into account:

- Elasticity of blood analog fluid;
- Constant inlet velocity (0.149m/s) – $\mathrm{Re} = \frac{\rho V_0 H}{\mu} = 180$;
- Steady state.



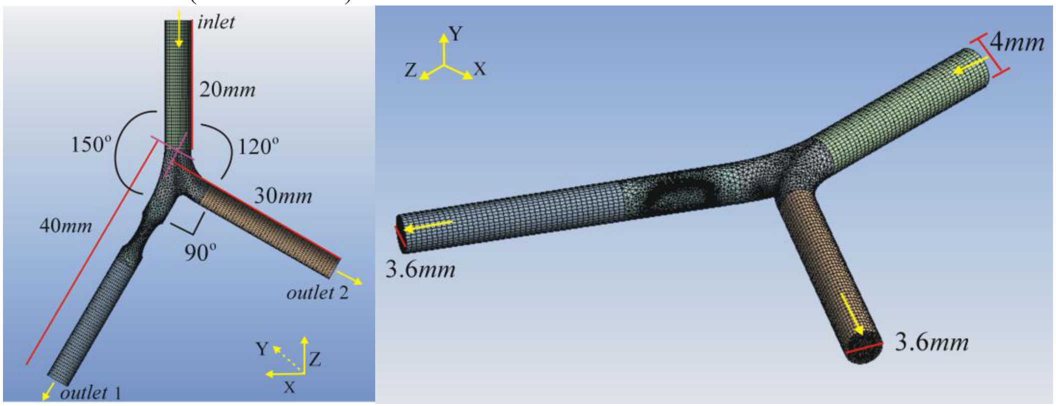
(a) (b)

Fig.1.(a) Geometry, dimensions and computational grid for an artery with a stenosis; (b) Detail of the computational grid - XY plane slice.

Figure 2 represents the geometry, dimensions and computational grid for a bifurcation with a stenosis (simplified representation of the left coronary artery). The geometry is composed by a main artery divided into two branches: one with a stenosis and another healthy. The domain for simulations was divided into five regions and different grids were created in each region. Figure 2a represents the XZ plane slice and Figure 2b the 3D representation. The following conditions were considered for the bifurcation with a stenosis:

- Shear-thinning of blood analog fluid:

- Pulsatile flow (transient state)



(a) (b)

Fig.2. Geometry, dimensions and computational grid for a bifurcation with a stenosis (simplified representation of a left coronary artery with a stenosis) (a) XZ plane slice; (b) 3D representation

2.2. Equations

The constitutive equation, which describes the elasticity of the blood analog fluid, has to be solved simultaneously with the conservative equations, which also depend on the elasticity.
The flow is described by:

$$\nabla\cdot(U)=0 \tag{1}$$

$$\frac{\partial(\rho U)}{\partial t}+\nabla\cdot(\rho UU)=-\nabla p+\nabla\cdot\tau_S+\nabla\cdot\tau_P \tag{2}$$

where U is the velocity, ρ the mean density, τ_S the stress tensor of the solvent part defined by:

$$\tau_S=2\eta_S D \tag{3}$$

where τ_S is the viscosity of the solvent part and D the deformation rate tensor defined by:

$$D=\frac{1}{2}\left(\nabla U+[\nabla U]^T\right) \tag{4}$$

The stress tensor of the polymeric part - τ_P of equation (2) - is defined by a constitutive equation. There are several models with different constitutive equations. The Phan-Thien-Tanner (PTT) linear model was chosen:

$$\left(1+\frac{\varepsilon\lambda}{\eta_P}tr(\tau_P)\right)\tau_P+\lambda\overset{\nabla}{\tau}_P=2\eta_P D \tag{5}$$

$\overset{\nabla}{\tau}_P$ is the Gordon-Schowalter derivative defined by:

$$\overset{\nabla}{\tau}_P=\frac{D}{Dt}\tau_P-\left[\nabla U^T\cdot\tau_P\right]-\left[\tau_P\cdot\nabla U\right]+\xi\left(\tau_P\cdot D+D\cdot\tau_P\right) \tag{6}$$

The parameters for two most common blood analog fluids, xanthan gum (XG) and polyacrylamide (PAA), are well known and described by Sousa et al. [8].

The PTT linear model was implemented in OpenFoam by Favero [9]. However, authors of the present paper are modifying the model to account for shear-thinning, simultaneously with elasticity – Modified PTT linear model.

Moreover, authors of the present paper implemented the pulsatile blood flow (pulsatile flow in the left coronary artery [10]) in the numerical code.

The inlet velocity profile (V_Z^{in}) depends on the mean inlet velocity (V_m^{in}) and on the position inside the artery (r):

$$V_Z^{in} = V_m^{in}\left(\frac{3n+1}{n+1}\right)\left[1-\left(\frac{r}{R}\right)^{\frac{n+1}{n}}\right] \tag{7}$$

$$V_m^{in} = a_0 + \sum_{i=1}^{8}\left[a_i \cos(itw) + b_i \sin(itw)\right] \tag{8}$$

V_m^{in} depends on the instant of the cardiac cycle (t). R is the ratio of the artery and n the parameter of the shear-thinning model. Parameters of Carreau Model for blood was used to account for shear-thinning [11].

3. Results

3.1. Artery with Stenosis

Results were obtained for a simpler geometry, an artery with a stenosis (without bifurcation) – Figure 1. Elasticity of blood analog fluid was considered, but shear-thinning of blood analog and pulsatile flow were not considered at this step.

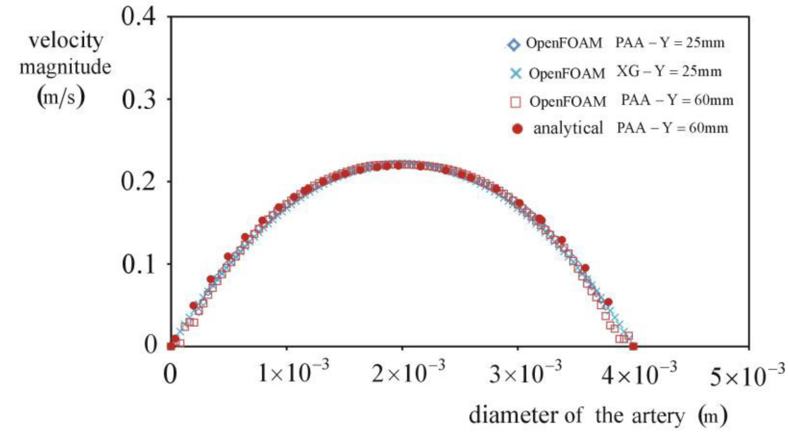
Figure 3 represents the velocity magnitude of PAA and XG along the diameter of the artery (X). While Figure 3a does not consider a stenosis, Figure 3b considers a stenosis. Velocity profiles were taken along the length of the artery: profile in the stenosis (Y=25mm) and after the stenosis (Y=60mm).

The velocity profile, after the stenosis (when the flow is stabilized), taken from numerical results (OpenFOAM code) is coincident to that obtained from the analytical solution (Oliveira and Pinho, 1998 [12]). In the stenosis, where De number is higher, the velocity profile is contracted and the maximum velocity is increased. Since De number considering PAA fluid is 10 times higher than considering XG fluid, the maximum of the velocity profile of PAA is higher than the maximum of the velocity profile of XG. The elasticity of the blood analog fluid leads to a more elongated velocity profile, increasing the maximum velocity.

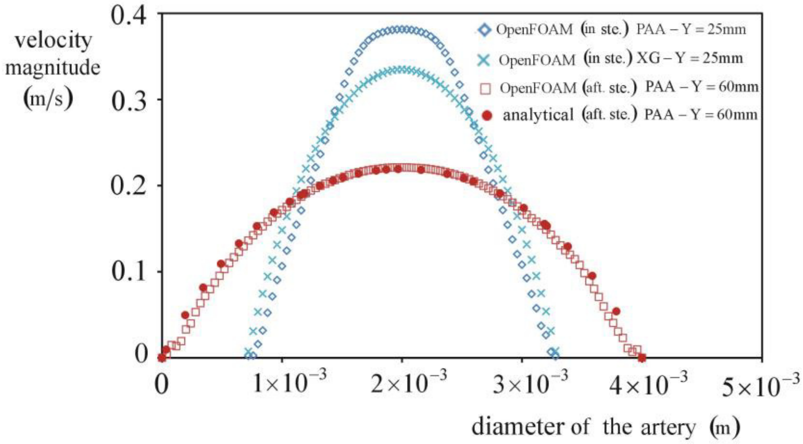
3.2. Bifurcation with Stenosis

One the other hand, results considering a bifurcation with a stenosis, pulsatile flow (transient state) and shear-thinning of blood analog fluid were obtained – Figure 4. Elasticity of the fluid was not considered.

Figure 4 represents the velocity profiles before the stenosis (Figure 4a), in the stenosis (Figure 4b) and after the stenosis (Figure 4c). Each figure shows the velocity profiles for three instants of the cardiac cycle (0.28s, 0.54s, 0.70s). Results obtained through OpenFOAM programming were also compared with those obtained through ANSYS package.

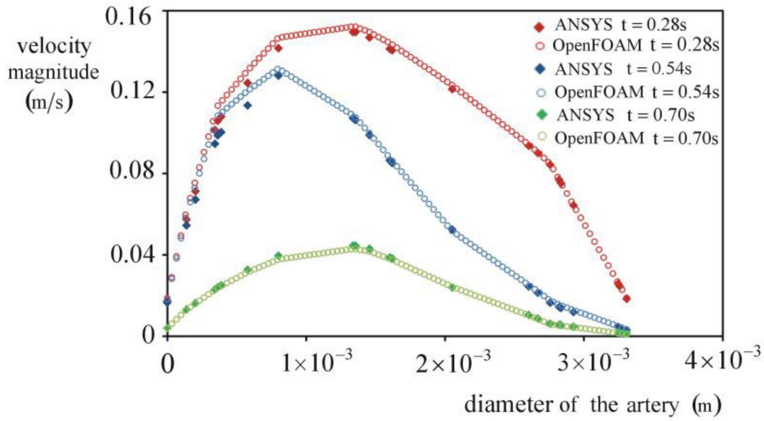


(a)

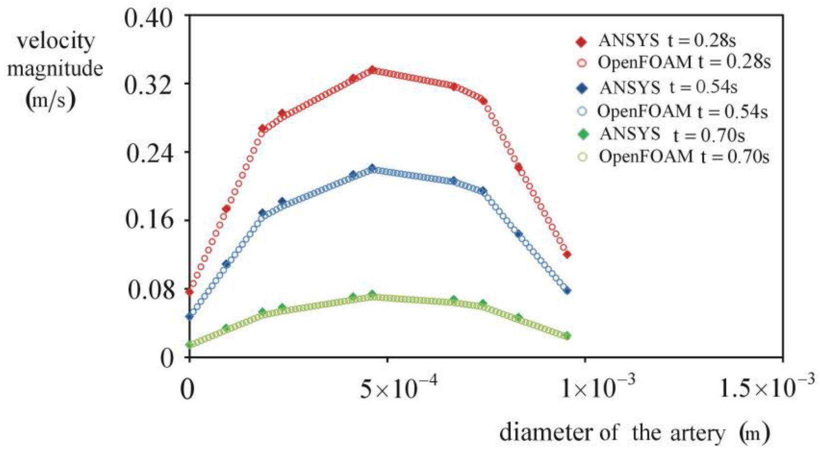


(b)

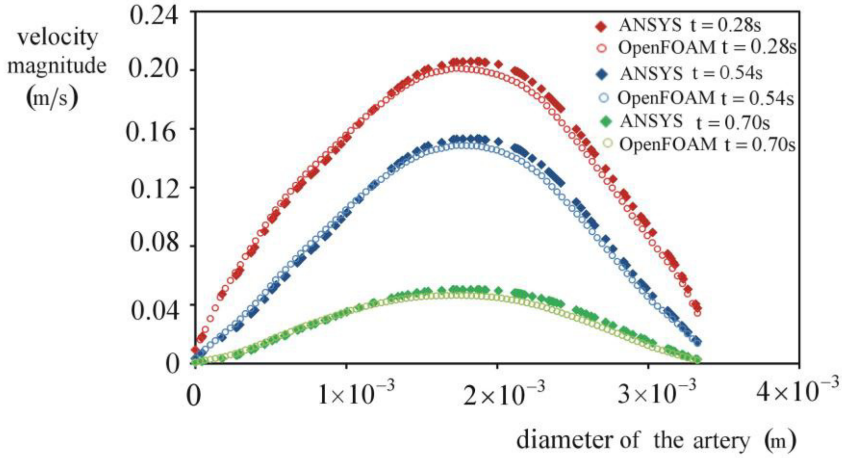
Fig. 3. Velocity magnitude along X - diameter of the artery (4mm) – (a) without stenosis (Deborah number ($\mathrm{De} = \frac{\lambda V_m}{R}$) equal to 2.81 for PAA and 0.281 for XG); (b) with a stenosis ($R_{stenosis}$ = 1.25 mm and De number in the stenosis equal to 4.53 for PAA and 0.432 for XG).



(a)



(b)



(c)

Fig. 4. Velocity magnitude vs. diameter of the artery, for three instants of the cardiac cycle (a) before the stenosis; (b) in the stenosis; (c) after the stenosis. Comparison between using OpenFOAM and ANSYS

The velocity profiles obtained by OpenFOAM are in agreement with those obtained by ANSYS. The velocity profile evolves during the cardiac cycle. Additionally, the profile is influenced by the geometry of the vessel. Before the stenosis, the maximum of the velocity profile is distorted to the side of the bifurcation. In and after the stenosis, the maximum of the velocity profile is in the middle of the artery.

4. Conclusions

To study the influence of the elasticity of the blood analog fluid, results obtained through OpenFOAM programming were compared to an analytical solution. The velocity profiles are in agreement. In the stenosis, the elasticity of the blood analog fluid leads to a more elongated velocity profile, increasing the maximum velocity.

To study the influence of the shear-thinning of the blood analog fluid and of the pulsatile flow, results obtained through OpenFOAM were compared with those obtained through ANSYS. The velocity profiles are in agreement. Before the stenosis, the maximum of the velocity is distorted to the side of the bifurcation.

In a future work, the blood flow in the left coronary artery with stenosis, considering, all simultaneously, elasticity of blood, shear-thinning of blood and pulsatile flow will be studied. The problem of convergence of the numerical code, taking into account all these considerations, simultaneously, should be solved step by step. The present work reports one of the steps.

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