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NUMERICAL STUDY OF BLOOD FLOW IN A BIFURCATION WITH A STENOSIS: PULSATILE FLOW AND ELASTICITY OF THE FLUID

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ABSTRACT

The present work reports a numerical study of the blood flow in a bifurcation with a stenosis using the CFD package OpenFOAM - an open source code freely modified. Pulsatile flow and elasticity of the fluid were considered and implemented in the numerical code. The velocity profiles taken from OpenFoam are in agreement with those obtained through an analytical solution (considering steady state and elasticity of the fluid) and also with the profiles obtained with ANSYS code (considering pulsatile flow and shear-thinning of the fluid).

Keywords: pulsatile flow, viscoelasticity, stenosis, bifurcation, OpenFOAM.

INTRODUCTION

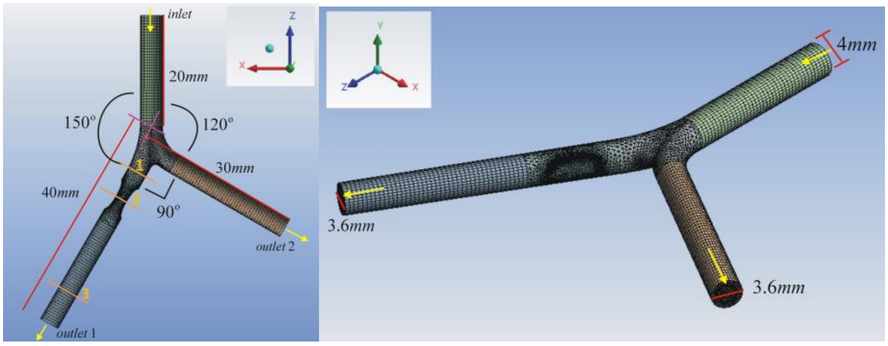
Cardiovascular diseases, caused by stenoses or aneurysms, are one of the main causes of death worldwide – Murray and Lopez (1997). The study of human blood flow has gained great importance and can contribute, as an auxiliary tool, to the prevention and treatment of such diseases. Several authors have been studying, numerically, perturbations in human blood flow: Botar et al. (2010) in the portal vein, Scotti et al. (2008) in an aortic aneurysms, Banerjee et al. (2008) in stenotic arteries and Sousa and Castro et al. (2011, 2012) in the carotid bifurcation.

As far as we know, the study of blood flow in a bifurcation with a stenosis is not a topic well explored in the literature. This topic is important since stenosis frequently occurs after the main bifurcation of the left coronary artery, and it can be a cause of death. Authors of the present paper started to study blood flow in a bifurcation with a stenosis (designed in CAD), through numerical methods using ANSYS code – Pinto et al. (2012). In that study, steady state and shear-thinning behavior of blood were considered.

In the present work, the authors intend to simulate blood flow considering pulsatile flow and the elasticity of the fluid. The CFD package OpenFOAM was used since it allows the direct access to the source code in order to control the specifications of the flow and the elasticity equations.

COMPUTATIONAL MESH AND EQUATIONS

The geometry was created in *SolidWorks* and imported to *ANSYS* to create the mesh. Afterwards, the mesh was imported to *OpenFOAM* to perform numerical simulations The domain was divided into five regions and different grids were created in each region – Figure 1. The “Tetrahedrons path conforming method” (tetrahedral mesh geometry) was used to create the grid in the bifurcation and stenosis regions while a simpler method, "Sweep method" (triangular prismatic mesh geometry) was used for the others regions.



a) b)

Fig.1 Geometry, dimensions and computational grid for the bifurcation with a stenosis
a) XZ plane slice, b) 3D representation

The constitutive equation, describing 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:

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

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

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

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

where τ_S is the viscosity of the solvent 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 through the Phan-Thien-Tanner (PTT) linear model:

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

Where $\overset{\nabla}{\tau}_P$ 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 of the most common blood analog fluids, xanthan gum (XG) and polyacrylamide (PAA), for PTT model, are well known and described by Sousa et. al. (2011) – Table 1.

Table 1 Parameters of xanthan gum (XG) and polyacrylamide (PAA) for PTT model (reference temperature 293.2 K).

Parameters	PAA	XG
ε	0.01	0.05
η_P (Pa.s)	0.498	0.05
η_S (Pa.s)	0.002	0.0015
λ (ms)	38.0	3.77
ξ	0	0

The PTT linear model was implemented in the OpenFoam by Favero (2009). However, the authors of the present paper are modifying the code to account for shear-thinning, effects simultaneously with elasticity effects – Modified PTT linear model.

Moreover, the authors of the present paper implemented the pulsatile blood flow (pulsatile flow in the left coronary artery (Ku, 1997)) 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):

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

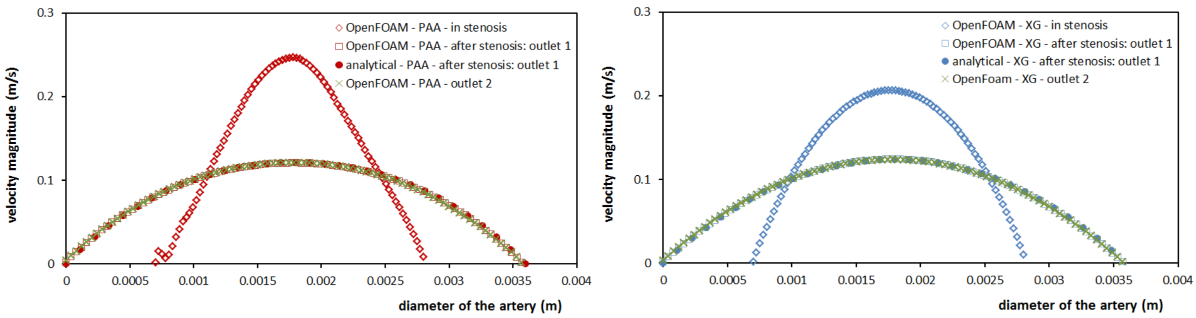
$$V_m^{in} = a_0 + \sum_{i=1}^{8}\left[a_i cos(itw) + b_i sin(itw)\right] \tag{8}$$

where V_m^{in} depends on the instant of the cardiac cycle (t), R is the radius of the artery and n the parameter of the shear-thinning model. Parameters of Carreau Model for blood were used to account for shear-thinning (Yilmaz and Gundogdu, 2008).

The influence of the stenosis resistance in the total resistance was evaluated (see Appendix A). The resistance of the stenosis has a very low contribution in the total resistance of the branch (0.3%). The resistance of the stenosis can be neglected and the flow rate at the outlet of each branch can be taken as 50% of the total inlet flow rate (see Appendix A). The velocity at outlet 2 was imposed according to this assumption.

RESULTS AND CONCLUSIONS

The velocity magnitude profiles vs. the diameter of the artery are represented in Figure 2 and 3. Figure 2 is for a position inside and after the stenosis (line 2 and 3 of Figure 1a) considering steady state and elasticity of the blood analog: polyacrylamide (PAA) with a Deborah number (De) in the stenosis of 1.72 (Figure 2a) and xanthan gum (XG) with Deborah number in the stenosis of 0.174 (Figure 2b). Figure 3 is for a position before the stenosis (line 1 of Figure 1a) considering pulsatile flow and shear-thinning behavior of the blood.



a) b)

Fig.2 Velocity magnitude profile vs. diameter of the artery inside and after the stenosis: steady state and elasticity of a) PAA fluid b) XG fluid.

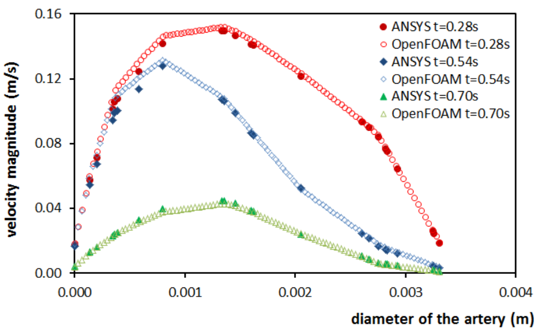


Fig.3 Velocity magnitude profile vs. diameter of the artery in a position before the stenosis: pulsatile flow and shear-thinning of blood.

After the stenosis (Figure 2a and 2b), the profiles taken from OpenFOAM are in agreement with the analytical solution obtained by Oliveira and Pinho (1999). The resistance in the stenosis was neglected comparatively to the total resistance (see Appendix A) .and so the velocity profiles at outlet sections 1 and 2 were taken equal.
In the stenosis, the maximum velocity profile is higher when PAA fluid is taken as blood analog. For PAA fluid, the Deborah number is 10 times higher than for XG fluid.
Before the stenosis (Figure 3), the velocity profiles obtained by OpenFOAM are in agreement with those obtained by ANSYS, for the three instants of the cardiac cycle. Moreover, the velocity profiles are asymmetric and the maximum is to the side of the bifurcation.
Results taking into account, bifurcation with stenosis, pulsatile flow, elasticity of the fluid and shear-thinning of the fluid, all the effects simultaneously, are still under development.

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APPENDIX A

Importance of the stenosis resistance in the total resistance

Figure A.1 represents the electric analog of the network vessels of the left coronary artery : P_0 is the pressure at the inlet of the left coronary artery and P_2 the pressure at the outlet (Kassab et al., 1997, Liu and Kassab, 2007, VanBavel and Spaan, 2012); R is the resistance and Q the flow rate (Sirca, 2008).

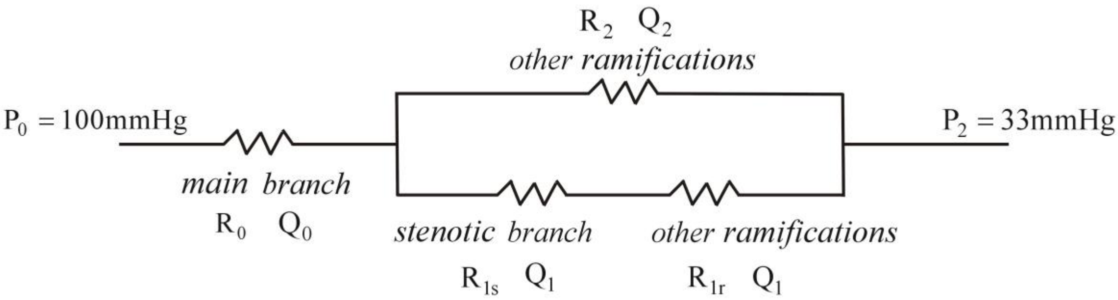

Fig.A.1 Schematic representation of the electric analog of the network vessels of the left coronary artery.

The following relationship is known as Murray´s Law (van der Giessen et al., 2011):

$$\frac{Q_1}{Q_2} = \left(\frac{D_1}{D_2}\right)^3 \tag{A.1}$$

where Q_1 and D_1 are the flow rate and the diameter, respectively, of the stenotic branch; and Q_2 and D_2 are the flow rate and the diameter of the healthy branch. D_1 and D_2 were considered equal, so, Q_1 and Q_2 are both 50% of the inlet flow rate, Q_0.

The resistance R_2 is defined as:

$$R_2 = \frac{P_0 - P_2}{Q_2} \tag{A.2}$$

and its value is 9.53×10^9 kg.m^{-4}.s^{-1}.

The total resistance $R_{1s} + R_{1r}$ is defined by:

$$R_{1s} + R_{1r} = \frac{P_0 - P_2}{Q_1} \tag{A.3}$$

and its value is also equal to 9.53×10^9 kg.m^{-4}.s^{-1}.

R_{1s} is defined as:

$$R_{1s} = \frac{(\Delta P)_{stenosis}}{Q_1} \qquad (A.4)$$

and its value is equal to 3.31×10^{7} kg.m^{-4}.s^{-1}, for a severe stenosis. R_{1r} is equal to 9.50×10^{9} kg.m^{-4}.s^{-1}.

The percentage of the stenosis resistance in the total resistance of the branch $\left(\frac{R_{1s}}{R_{1s}+R_{1r}}\right)$ is, applying the correlations above described,0.3%. The resistance of the stenosis can then be neglected.

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