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IMPORTANCE OF BLOOD ELASTIC PROPERTY IN THE HEMODYNAMICS AT MACRO-SCALE - NUMERICAL STUDY

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ABSTRACT

The present numerical work is about the importance of blood elasticity in the hemodynamics at the macro-scale. The flow in a stenotic bifurcated artery was used as example. The Multi-mode Phan-Thien-Tanner (PTT) model was implemented in OpenFOAM® code to take into account the blood elasticity. In locations where shear rates are higher than 10 s^{-1}, differences in the velocity field are not observed, considering Multi-mode PTT model (shear-thinning and elastic behaviors of blood) or Carreau model (only shear-thinning behavior). The blood elasticity effect is insignificant. Artery locations where shear rates are lower than 10 s^{-1}(near the bifurcation and stenosis) the blood elasticity has an important role on the blood flow pattern.

Keywords: blood elastic property, hemodynamic, macro-scale, multi-mode PTT model, OpenFOAM®.

INTRODUCTION

Cerebrovascular and cardiovascular diseases are nowadays the main cause of human mortality in advanced countries (Mozaffarian et al. 2015). From clinical practice, specific sites in human circulatory system are sensitive to the development of coronary diseases such as atherosclerosis. Numerical studies of human blood flow have gained importance and can contribute, as an auxiliary tool, to the prevention and treatment of such diseases. However, none of the numerical studies in the literature (Alastruey et al. 2011, van der Giessen et al. 2011, Chaichana et al. 2013) considers, simultaneously, three realistic approaches: shear-thinning and elastic rheological blood behaviors and pulsatile flow. The goal of the present work is to conclude about the importance of blood elasticity in hemodynamic flow patterns at macro-scale arteries. For this purpose, the Multi-mode PTT model was implemented in the open source code OpenFOAM®.

NUMERICAL MODEL

A bifurcated stenotic artery model is represented in Fig. 1 (middle plane of the artery). The geometric dimensions and the flow direction (V_y^*) are observable in the figure.

The model was created in SolidWorks® and imported to Fluent Meshing ANSYS® to generate a refined mesh. The mesh was then feed to OpenFOAM® for numerical simulations. In this open source code, the constitutive equation, describing the blood rheology, was implemented and à posteriori solved simultaneously with the conservative equations.

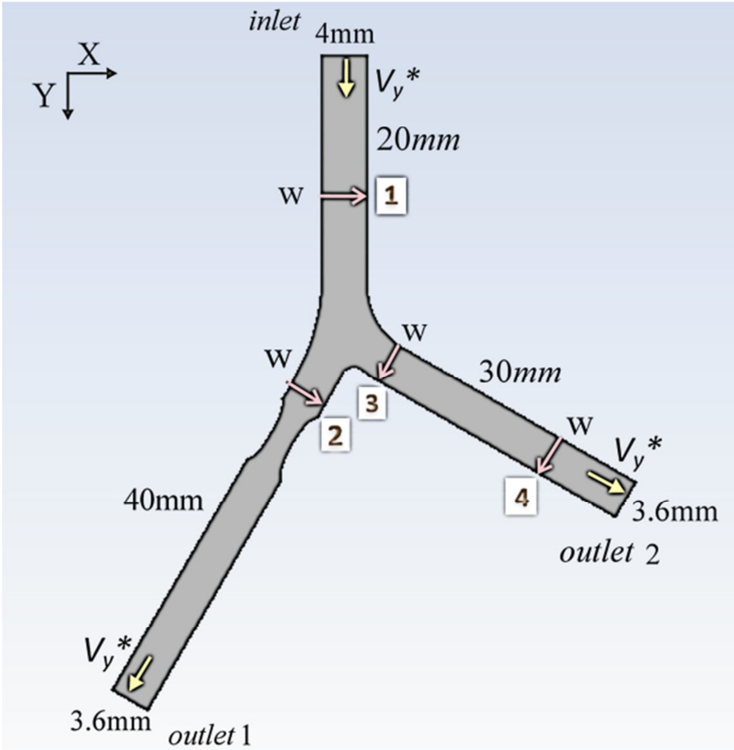

Fig. 1 - Stenotic bifurcated artery (middle plane representation).

The mass and momentum conservative equations are:

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

$$\frac{\partial(\rho \mathbf{U})}{\partial t} + \nabla \cdot (\rho \mathbf{U}.\mathbf{U}) = -\nabla \mathbf{p} + \nabla \cdot \boldsymbol{\tau}_\mathbf{S} + \nabla \cdot \boldsymbol{\tau}_\mathbf{P} \tag{2}$$

where $\mathbf{U}$ is the velocity vector, ρ the mean density and $\boldsymbol{\tau}_\mathbf{S}$ the stress tensor of the solvent part defined by:

$$\boldsymbol{\tau}_\mathbf{S} = 2\eta_S \mathbf{D} \tag{3}$$

where η_S is the solvent viscosity and $\mathbf{D}$ the deformation rate tensor:

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

The stress tensor of the polymeric part, $\boldsymbol{\tau}_\mathbf{P}$, is defined by a constitutive equation. Several constitutive equations, representing the elasticity of the blood (Favero, 2009), are cited in the literature. The one chosen and implemented in the OpenFOAM® code was the Multi-mode PTT model with the respective parameters at 37 ºC given by Campo-Deaño et al. (2013).

Pulsatile blood flow was considered. The mean inlet velocity (V_{in}) along the cardiac cycle is represented in Fig. 2. The systolic phase (maximum velocity in the cardiac cycle) and the diastole phase can be observed. Moreover, the instantaneous velocity profile was imposed fully developed.

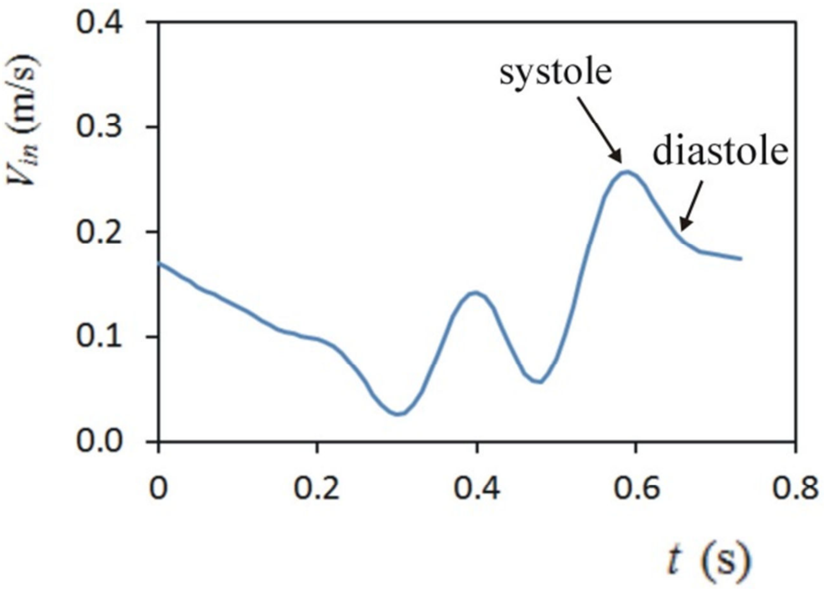

Fig. 2 - Mean inlet velocity along the cardiac cycle

The flow splitting in the bifurcation is according to Murray´s Law (Murray, 1926):

$$\frac{Q_1}{Q} = \left(\frac{d_1}{d}\right)^3 ; \frac{Q_2}{Q} = \left(\frac{d_2}{d}\right)^3 \quad (5)$$

$$Q = Q_1 + Q_2 \quad (6)$$

where Q and d are the flow rate and the diameter of the main artery, respectively. (Q_1, Q_2 and d_1, d_2 are the flow rates and diameters of each branched artery).

At the arteries wall, assumed rigid, the velocity was considered zero (no-slip condition).

RESULTS

Fig. 3 shows, for systolic phase, the velocity profiles, in the middle plane, along the blood flow direction (V_y^*), considering Multi-mode PTT and Carreau models. Velocity profiles imposing Carreau model (only shear-thinning blood behavior) are also represented for comparison with outlines using Multi-mode PTT (elastic and shear-thinning blood behaviors). Fig. 3a represents the profiles in the main branch of the artery, (LINE 1 of Fig. 1); Fig. 3b in the stenotic branch just after the bifurcation and before the stenosis (LINE 2 of Fig. 1); Fig. 3c in the non-stenotic branch just after the bifurcation (line LINE 3 of Fig. 1); and Fig. 3d in the non-stenotic branch far from the bifurcation (LINE 4 of Fig. 1).

Apparent blood Viscosity vs. Shear Rate is represented, in Fig. 4, for Carreau model (Yilmaz and Gundoglu et al. 2008), Multi-mode PPT model (Campo-Deaño et al. 2013), experimental data (Valant et al. 2016) and simulated data using Multi-mode PTT model implemented in OpenFOAM® code (present work). Carreau model fits well experimental data only for shear rate values higher than 10 s^{-1}, while Multi-mode PTT fits well blood experimental data for all the shear rate range.

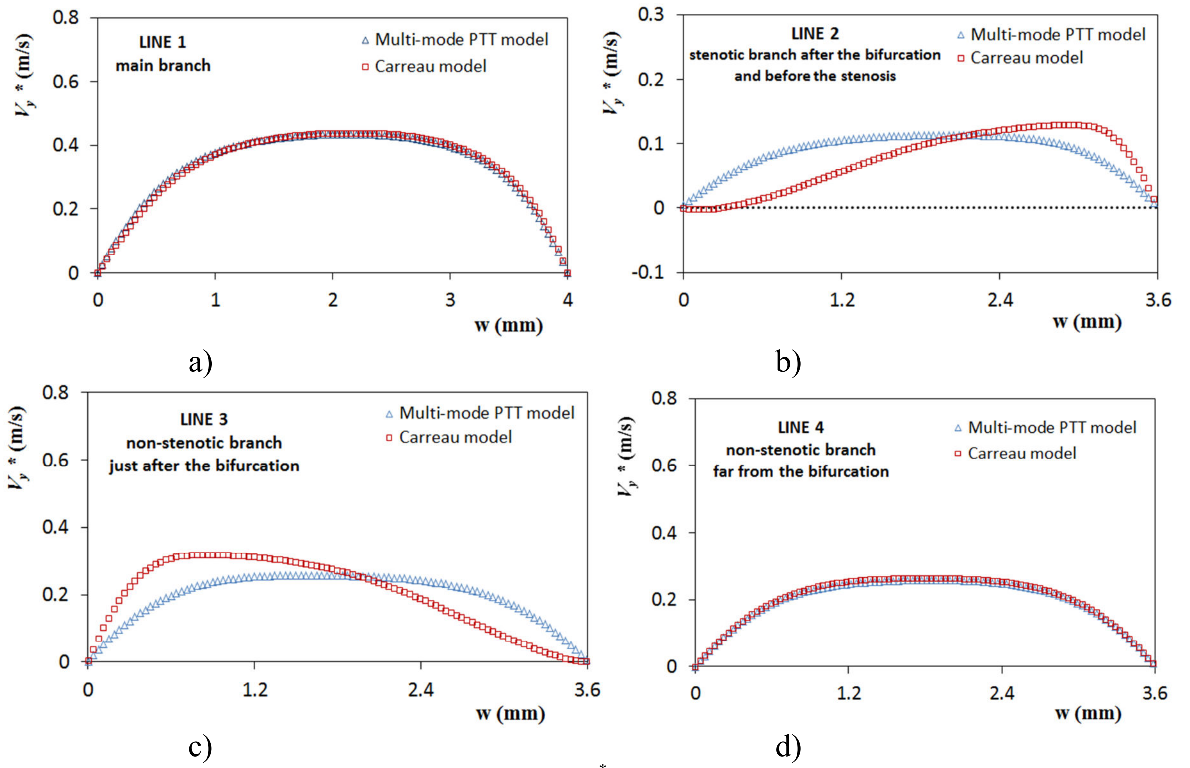

Fig. 3 - Velocity profiles along blood flow direction (V_y^*) in: a) the main branch (LINE 1); b) the stenotic branch after the bifurcation and before the stenosis (LINE 2); c) the non-stenotic branch just after the bifurcation (LINE 3); d) in the non-stenotic branch far from the bifurcation (LINE 4).

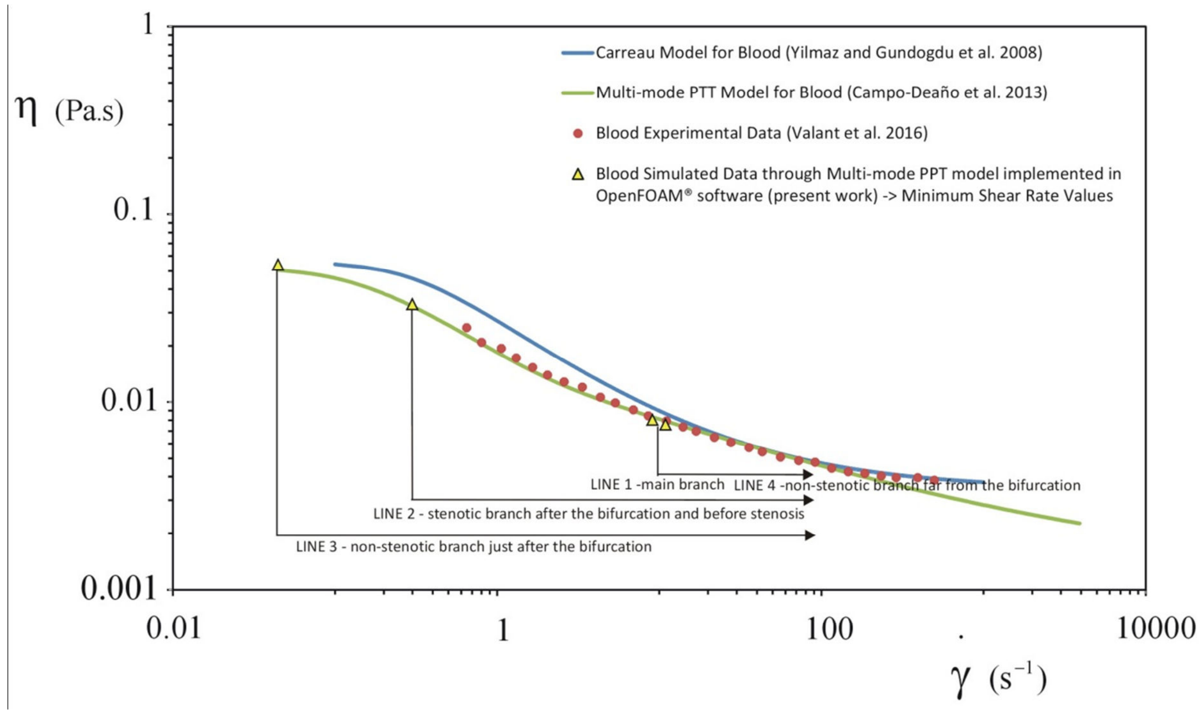

Fig. 4 - Apparent viscosity vs. Shear Rate using Carreau model for blood rheology (Yilmaz and Gundoglu et al. 2008), Multi-mode PPT model (Campo-Deaño et al. 2013), blood experimental data (Valant et al. 2016) and blood simulated data using Multi-mode PTT model implemented in OpenFOAM® code (present work).

CONCLUSIONS

In the main artery (LINE 1 - Fig. 3a) and in the non-stenotic branch far from the bifurcation (LINE 4 - Fig. 3d), very similar velocity profiles are observed considering Multi-mode PTT or Carreau models, during the systolic phase. Since the minimum shear rates observed in LINE 2 and LINE 4 (far from the bifurcation and from the stenosis) are 9.08 and 10.94 s^{-1}, the total shear rate range is inside the applicability range of Carreau model (see Fig. 4). Blood elastic effects are not relevant in these two positions of the artery.

In the stenotic branch just after the bifurcation and before the stenosis (LINE 2 - Fig. 3b) and in the non-stenotic branch also after the bifurcation (LINE 3 - Fig. 3c), dissimilarities are observed in the velocity profiles using Multi-mode PTT (symmetric velocity profile relatively to the center of the artery) and Carreau model (maximum velocity near the artery wall). Once the minimum shear rates achieved in LINE 2 and LINE 3 (near the bifurcation) are 0.30 and 0.045 s^{-1}, several values are inside the range where only Multi-mode PTT model is accurate (see Fig. 4). The blood elastic behavior is determinant in these regions.

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