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LES OF NON-NEWTONIAN PHYSIOLOGICAL BLOOD FLOW

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ABSTRACT

Large Eddy Simulation (LES) is performed to study the physiological pulsatile transition to turbulent non-Newtonian blood flow through a 3D model of arterial stenosis using the different non-Newtonian blood viscosity models. The computational domain has been chosen is a simple channel with a biological type stenosis formed eccentrically on the top wall. The physiological pulsation is generated at the inlet of the model using the fourth harmonic of the Fourier series of the physiological pressure pulse (Womersley [1]). The computational results are presented in terms of the post-stenotic re-circulation zone, shear stress, mean and turbulent kinetic energy.

Key Words: *non-Newtonian bloodflow, re-circulation zone, wall shear stress, stenosis, LES.*

1 INTRODUCTION

The term arterial stenosis refers the narrowing of an artery where the cross-sectional area of blood vessel reduces. Blood is a non-Newtonian incompressible viscoelastic fluid (Fung [2], pp.53). At shear rates above about $100\ \mathrm{s}^{-1}$, the blood viscosity tends towards an asymptotic value. If the shear rates fall below that asymptotic level, the viscosity of blood increases and the non-Newtonian properties of blood being exhibited (Berger and Jou [3]), especially when the shear rates drop below $10\ \mathrm{s}^{-1}$ (Huang *et al*. [4]).

Very few studies which are related to the non-Newtonian blood flow in arterial stenosis, such as Tu *et al*. [5], Buchanan *et al*. [6], Neofytou and Drikakis [7], Hron *et al*. [8] and Valencia and Villanueva [9] used different blood viscosity models, however, all these studies are conducted only for laminar flow. Most recently Paul *et al*. [10] have investigated the pulsatile turbulent blood flow through a model of arterial stenosis applying the LES technique. Using the first harmonic of the Fourier series of the pressure pulse, a Large-eddy simulation of the physiological pulsatile flow in the same model is performed by Molla *et al*. [11]. In these papers, the investigation has been done assuming that the blood is a Newtonian fluid. However, the recent investigation shows that the global maximum shear rate during some periods of pulsation receives a result which is less than the range of the non-Newtonian shear rate ($100\ \mathrm{s}^{-1}$). Therefore, it would be quite reasonable to account in the computation the blood as a non-Newtonian fluid to get more accurate insight of the transition of the blood flow through the stenosis. In this regard, various blood viscosity models are applied and their effects are examined in the paper.

The geometry chosen in the simulation (Fig. 1(a)) consists of a 3D channel with one sided cosine shape stenosis on the upper wall centred at $y/L = 0.0$ with a 50 cross-sectional area reduction at the centre. In Fig. 1(b)-(c), the inlet velocity profile is presented for one pulsation for the Reynolds number of 2000 and the Womersley number of 10.5. Note that the velocity in frame (b) is recorded at very close to the bottom wall. In the simulation, no slip boundary conditions are used for both the lower and upper walls

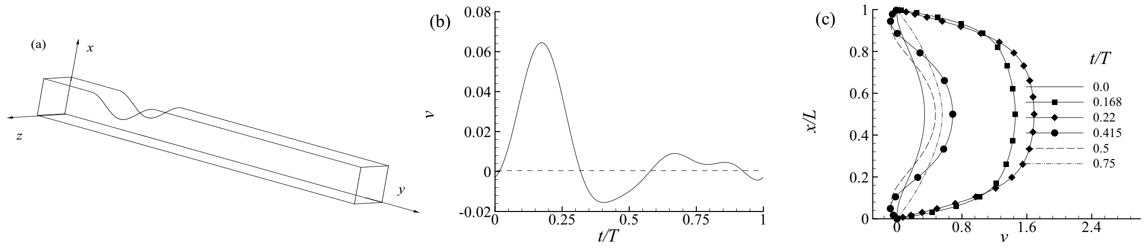

Figure 1: (a) A schematic of the model with coordinate system and (b-c) Streamwise inlet velocity.

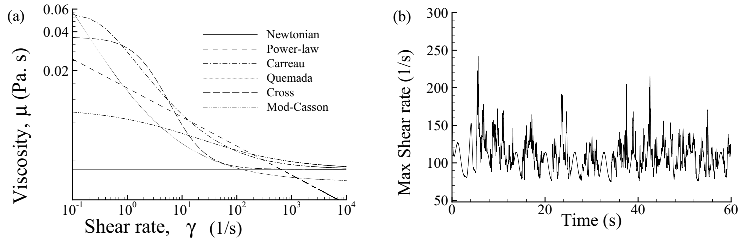

Figure 2: (a) Relation between the apparent shear rate and viscosity for the different models (b) Global maximum shear rate $|\dot{\gamma}|$ against time for the Power-law model

of the model, and a convective boundary condition at the outlet For the spanwise boundaries, periodic boundary conditions are applied for modelling the spanwise homogeneous flow.

The LES code is written based on the finite volume method with collocated grid arrangement which is second order accurate in space and time. For the present computation the grid arrangement is taken as $50 \times 200 \times 50$ along the cross-stream, streamwise and spanwise direction, respectively, with a constant timestep of 10^{-3}. The sensitivity tests on grid and timestep are performed and the above grid arrangment is sufficient to resolve the large scale flows. In addition, the code is validated with suitable experimental data, the details are available in Molla [12].

2 RESULTS AND DISCUSSION

The relation between the apparent shear rates and the viscosity for the five non-Newtonian blood viscosity models along with the Newtonian one is presented in Fig. 2(a). From this figure, it is seen that for low shear rates (e.g. $< 100\,\mathrm{s}^{-1}$) the non-Newtonian blood viscosity is higher than that of the Newtonian model. Moreover, the necessity of using the non-Newtonian model is very much clear by observing the range of the global maximum shear rate in Fig. 2(b) for the Power-law model. Similar distributions of the shear-rate are found for the other models.

Fig. 3(a-f) depict the post-stenotic re-circulation zones in terms of the mean streamlines for the different models. The length of the re-circulation region is enlarged in the non-Newtonian models, which is an alarming condition at the pathological point of view since the blood in the re-circulation region is re-circulated for a long time and is stagnant in this region that could cause the blood clot or thrombosis.

The mean shear stress, $\tau_{xy}/\rho\bar{V}^2$, distributions are plotted in Fig. 4(a-b) respectively at the upper wall and lower wall. At the upper wall the stress drop is higher in the cases of non-Newtonian model than that of the Newtonian model and the maximum stress drop is found for the Power-law model. The magnitude of this stress drops to -0.07730 which is about 32 higher than the Newtonian model for which it is $-0.058\ 9$. Interestingly, the stress drop for all the models is occurring at a same streamwise

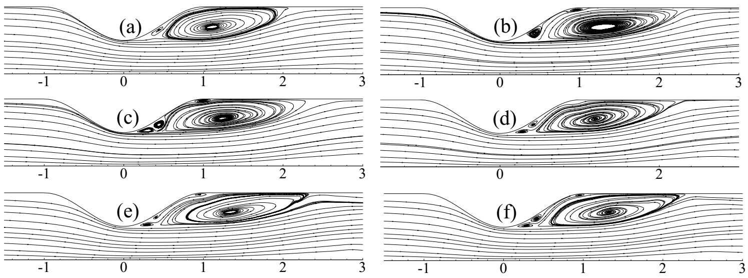

Figure 3: Post-stenotic recirculation zone at (a) Newtonian (b) Power-law (c) Carreau (d) Quemada (e) Cross and (f) Modified-Casson models.

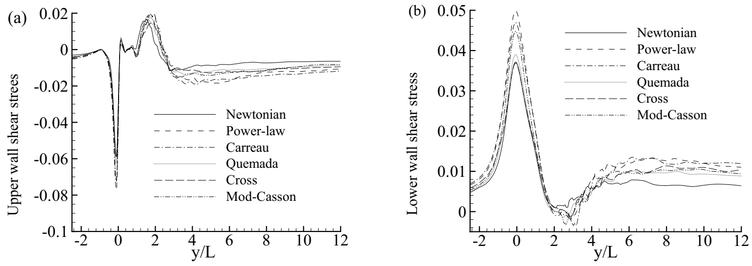

Figure 4: Shear stress at the (a) upper wall and (b) lower wall for the different blood viscosity models.

location, $y/L = -0.12505$. The difference between the non-Newtonian and Newtonian models of the shear distribution is distinguishable in the post-stenosis region, however, in the laminar region a small difference is found. Moreover, in the further downstream region, the upper wall shear stresses for all the non-Newtonian models are always smaller than the case of Newtonian model. On the other hand, the maximum shear stress at the lower wall occurs at $y/L = -0.059$ for all the models, but the most largest value is found again for the Power-law model which is 0.04989. In contrast to the upper wall, the non-Newtonian models give higher shear stress at the lower wall in the further downstream region of the stenosis.

The effects of various non-Newtonian models on the mean and turbulent kinetic energy are illustrated in Fig. 5. The mean kinetic energy in the turbulent region ($1.0 < y/L <$ .0) varies in the non-Newtonian models with the magnitude that is slightly higher in the Carreau and Quemada models compared to the Newtonian model. However, the curves are identical at the upstream of the stenosis. The significant effects are reported on the results of the turbulent kinetic energy in frame (b). The peak in the post-lip region ($1.0 < y/L < 3.0$) occurs in the Newtonian model, while all the non-Newtonian models produce higher turbulent kinetic energy in the further downstream because of the fact that the physiological oscillation is reduced by the higher viscosity in the non-Newtonian models which causes delay in the transition process.

3 CONCLUSIONS

Non-Newtonian physiological flow in the model of arterial stenosis has been investigated by using the LES technique. The global maximum shear rate for the different viscosity models falls below the non-Newtonian range of 100 s^{-1} which clearly indicated the necessity of applying the various non-Newtonian blood viscosity models in the investigation. The post-stenotic re-circulation region is extended slightly in the non-Newtonian models, in the pathological point of view, this usually increases the possibility of thrombosis or blood clot. The shear stress results are influenced by the non-Newtonian

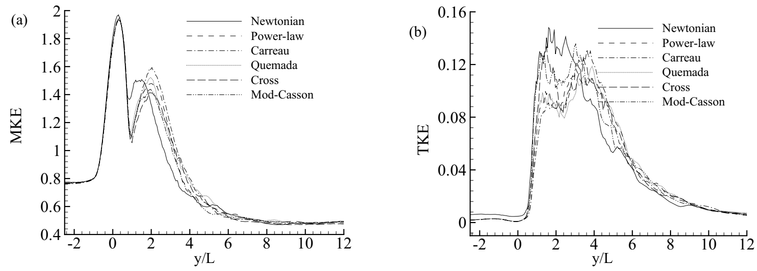

Figure 5: (a) Mean and (b) turbulent kinetic energy for the different blood viscosity models.

models producing the maximum rise at the lower wall while dropping significantly at the upper wall. The intensity of the turbulent kinetic energy is also affected by the choice of the blood viscosity models. Despite the simplicity in the vessel model, the LES results of the non-Newtonian blood flow presented in the paper would provide a better insight in the understanding of the flow transition and turbulent downstream of a real stenosed artery and the formation of atherosclerosis. The natural extension of this work is to consider a more biological realistic model, e.g. circular and flexible artery and apply the LES to study the transition of the non-Newtonian blood flow.

References

[1] J. R. Womersley, Method for the calculation of velocity, rate of flow and viscous drag in arteries when the pressure gradient is known, *J. Physiology*, 155 (1955) 553–563.

[2] Y. C. Fung, Biomechanics: Circulation, 2nd edition, Springer,1997.

[3] S. A. Berger, L.-D. Jou, Flows in stenotic vessels, *Annu. Rev. Fluid Mech.*, 32 (2000) 347–382.

[4] C. R. Huang, W. D. Pan, H. W. Chen, A. L. Copley, Thixotropic properties of whole blood from healty human subjects, *Biorheology* , 24 (1987) 795–801.

[5] C. Tu, M. Delville, Pulsatile flow of non-Newtonian fluids through arterial stenosis, *J. Biomechanics*, 29 (7) (1996) 899–908.

[6] J. R. B. Jr, C. Kleinstreuer, J. K. Comer, Rheological effects on pulsatile hemodynamics in a stenose tube., *Computers and Fluids*, 29 (2000) 695–724.

[7] P. Neofytou, D. Drikakis, Effects of blood models on flows through a stenosis, *Int. J. Numer. Meth. Fluids*, 43 (2003) 597–635.

[8] J. Hron, J. Malek, S. Turek, A numerical investigation of flows shear-thinning fluids with applications to blood rheology, *Int. J. Numerical Methods in Fluids* , 32 (2000) 863–879.

[9] A. Valencia, M. Villanueva, Unsteady flow and mass transfer in models of stenotic arteries considering fluid-structure interaction, *Int. Communication Heat and Mass Transfer*, 33 (2006) 966–975.

[10] M. C. Paul, M. M. Molla, G. Roditi, Large-eddy simulation of pulsatile blood flow, *Medical Engineering and Physics*, 31 (2009) 153–159.

[11] M. M. Molla, M. C. Paul, G. Roditi, Physiological flow in a model of arterial stenosis, *J. Biomechanics*, 41(S1) (2008) S243.

[12] M. M. Molla, Large eddy simulation of pulsatile flow in the models of arterial stenosis and aneurysm, PhD thesis (in preparation) 2009, University of Glasgow, UK.