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Difference Between Newtonion And Non Newtonian Models Of Blood Flow Through Artery

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Abstract: Pulsatile blood flow in an aorta of normal subject is studied by Computational Fluid Dynamics (CFD) simulations. The main intention of this study is to determine the influence of the non-Newtonian nature of blood on a pulsatile flow through an aorta. The usual Newtonian model of blood viscosity and a non- Newtonian blood model are used to study the velocity distributions, wall pressure and wall shear stress in the aorta over the entire cardiac cycle. Realistic boundary conditions are applied at various branches of the aorta model. The difference between non-Newtonian and Newtonian blood flow models is investigated at four different time instants in the fifth cardiac cycle. This study revealed that, the overall velocity distributions and wall pressure distributions of the aorta for a non-Newtonian fluid model are similar to the same obtained from Newtonian fluid model but the non-Newtonian nature of blood caused a considerable increase in Wall Shear Stress (WSS) value. The maximum wall shear stress value in the aorta for Newtonian fluid model was 241.706 Pa and for non-Newtonian fluid model was 249.827 Pa. Based on the results; it is observed that the non-Newtonian nature of blood affects WSS value. Therefore, it is concluded that the non-Newtonian flow model for blood has to be considered for the flow simulation in aorta of normal subject.
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Introduction:

A numerical simulation to investigate the Non-Newtonian modeling effects on physiological flows in a three dimensional idealized stenosed carotid artery with 75% severity (by area) is taken from patient specific model.[1] The wall vessel is considered to be rigid. Oscillatory physiological and parabolic velocity profile has been imposed for inlet boundary condition.[2] Where the physiological waveform was performed using a Fourier series with sixteen harmonics. The investigation has a Reynolds number range of 94 to 1120. Low Reynolds number $k-\omega$ model is used as governing equation. The investigation has been carried out to characterize two Non-Newtonian constitutive equations of blood, namely, (i) Carreau and (ii) Cross models.[3] The Newtonian model has been investigated also to study the physics of fluid. The results of Newtonian model are compared with the Non-Newtonian models. The numerical results are presented in terms of pressure, wall shear stress distributions and the streamlines contours. At early systole pressure differences between Newtonian and Non-Newtonian models are observed at pre-stenotic, throat and immediately after throat regions.[4,5]

In the case of wall shear stress, some differences between Newtonian and Non-Newtonian models are observed when the flows are the minimum such as at early systole or diastole. It is known that blood is Bingham plastic fluid. So the viscosity of blood will decrease with increase in shear rate and when shear rate will be greater than 100 then viscosity will be constant.[6] The viscosity of Newtonian model is less than that of non-Newtonian model when shear rate is less than 100, but viscosity of all models is equal when shear rate is equal to 100. When Reynolds number is very low then pressure and WSS of Newtonian model will be less than that of non-Newtonian model but opposite scenario will be seen for velocity distribution. Since early systole and diastole there are comparatively low Reynolds numbers, the results of Newtonian and non-Newtonian condition may be different at early systole and diastole.[7] On the other hand maximum Reynolds number is seen at peak systole. So the results of Newtonian and non Newtonian condition follow each other at peak systole. Again the velocity of the throat region is high for any time instant. So the results of Newtonian and non Newtonian condition may be same at the throat region but different at the pre and post stenotic region. The results of pressure, WSS, and velocity distribution are discussed below with respective figure. [8,9]

The cardiovascular system maintains an adequate blood flow to all cells in the body. The flow of blood in the cardiovascular system depends upon the pumping mechanism of the heart. This mechanism induces the blood to flow in pulsatile nature. [10] The aorta is the largest and most important artery in cardiovascular system; it carries blood from the heart to other organs in a body. The occurrence of many diseases in cardiovascular system has been associated with blood flow behaviour in the blood vessels. The investigation on blood flow in aorta is crucial because the flow of blood changes hemodynamic stresses act upon the aortic wall. The relationship between hemodynamic stresses and corresponding changes in the layer of blood vessel wall is the major cause of aneurysm, other lesions. Therefore, the interaction between hemodynamic stresses and physiological behavior of blood vessel wall plays an important role in aneurysm formation, progression and rupture. Computational Fluid Dynamics (CFD) and Fluid-Structure Interaction (FSI) techniques have been widely used for simulating the blood flow in idealized and patient-specific aorta models. [11]

The blood is a non-Newtonian fluid and it follows Newtonian nature when the shear rate is above 100 s^{-1}. The effect of non-Newtonian behavior of flow is not significant in large blood vessels like aorta, where the shear rate is high. Considering the blood as a Newtonian fluid is a satisfactory assumption for large arteries such as aorta. In transient analysis, the non-Newtonian flow effects could become significant when the shear rate is below 100 s^{-1}. Some authors concluded that the non-Newtonian fluid approximation for flow in large arteries is crucial while others found it is an unimportant assumption. Gijsen et al. highlighted various differences between non-Newtonian and Newtonian flow patterns when they studied flow through 90° curved tube. Li et al. simulated blood flow through an Abdominal Aortic Aneurysm (AAA) model with Stent-Graft (SG) using non-Newtonian fluid assumption. They discussed about key biomechanical factors which are causing SG migration. The authors concluded that the blood flow conditions, aneurysm and SG geometries were major reasons for SG migration. Amblard et al. developed a methodology using non-Newtonian fluid approximation to observe the relation between the aorta's wall and endocraft to find when type I endoleaks could occur. They evaluated the stresses on the aorta's wall generated by the blood flow. [12]

In the present study, CFD approach is used to simulate the blood flow through an aorta with Newtonian fluid assumption for assessing the hemodynamic parameters. This study also depicts the influence of non-Newtonian nature of blood on various physiologically important flow parameters like velocity, wall pressure and wall shear stress (WSS). The Casson non-Newtonian model is used in this study because of its capability in representing the non-Newtonian blood rheology. The blood flow through a bifurcation with an aneurysm in the cerebrovascular system was simulated with the Casson non-Newtonian model by Perktold et al. Perktold et al. studied pulsatile flow characteristics through a human carotid bifurcation with the Casson model approximation. The blood assumed as power-law non-Newtonian fluid by Liepsch et al. for their flow study through renal artery and T-shaped bifurcations. The complex flow regions could happen in the abdominal aorta segment due to bifurcations, branches and curvature of the arteries. Wiwatanapataphee et al. studied the effect of branching vessel on the pulsatile blood flow in the human coronary artery with non-Newtonian fluid assumption. The authors concluded that the branching of artery influences the flow in the artery substantially. Shahcheraghi et al. described the influence of aortic branches on flow of blood. In this work, those aortic branches recommended by Shahcheraghi et al. are considered. The aorta model is simplified by excluding coronary arteries, intercostal arteries, gonadal artery, arteries branching from the brachiocephalic trunk, subclavian artery and celiac trunk. The inflow data to the aorta used in this study was measured at the ascending aorta past the coronary arteries and the coronary arteries carry approximately 4-5 percent of the whole cardiac output. Hence, the coronary arteries are not considered in the present work. The intercostal arteries carry less than 1percent of the whole cardiac output and hence these arteries are not considered in the present work. The brachiocephalic trunk, common carotid artery, left subclavian artery, celiac trunk, renal arteries, superior mesenteric artery, inferior mesenteric artery and common iliac arteries are included in the aorta model. In order to reduce the computational time the other arteries and branches are not considered in the aorta model. [13]

There are substantial studies reported on the CFD simulation of blood flow which investigated the influence of the non- Newtonian nature of blood on a pulsatile flow in idealized aorta, coronary artery and carotid artery. The CFD analyses have been carried out by a few authors with symmetric aortic bifurcation geometry. The aortic bifurcation is not a symmetric geometry and the influence of asymmetry in geometry on flow analysis is crucial. To the best of knowledge of authors, the CFD analysis of blood flow in an aorta model with Newtonian, non-Newtonian fluid assumptions had not been attempted. There are a few studies which have been conducted with realistic geometry of aorta. The purpose of this study is to find the influence of the non-Newtonian nature of blood on a pulsatile flow through an aorta. The physiologically important flow parameters such as velocity distribution, wall pressure and wall shear stress are estimated in the aorta through CFD simulation with both non-Newtonian and Newtonian fluid models. [14,15]

Newtonian model:

The pulsatile blood flow in aorta was investigated by using transient analysis. A time varying pulsatile velocity profile at the ascending aorta inlet and pressure waveforms at the outlet of iliac arteries were imposed in simulations based on data from Olufsen et al. The cardiac cycle period was 1 s with peak blood flow occurred at 0.14 s. A fifteen percent of the inlet flow volume was assumed for brachiocephalic artery; five percent of the inlet flow volume was prescribed at common carotid artery and left subclavian artery. [16] A total proportion of ten percent of the thoracic mass flow was considered for each renal artery and pulsatile pressure was fixed at other branches. Blood was treated as a homogeneous, Newtonian and an incompressible fluid. The blood flow can be considered as laminar in large blood vessels like aorta and it was found to be laminar in Abdominal Aortic Aneurysms (AAAs) during exercise. In this study, the maximum Reynolds number based on the flow velocity was 3971. Since the maximum Reynolds number was lower than the threshold Reynolds number the flow was assumed as laminar. Various cardiac cycles are required for achieving convergence for the transient analysis. The fifth cardiac cycle was used as the final periodic solution to obtain the hemodynamic parameters from the model. The other properties and boundary conditions were same as that of steady state analysis. [17]

Non-Newtonian model: Shear thinning or pseudoplastic fluids are the fluids whose effective viscosity decreases with increase in shear rate. This fluid structure is time-independent. The Casson model was recommended for shear thinning liquids. In the present work, the Casson model was used in transient simulation to approximate the non-Newtonian flow. The dynamic viscosity for the Casson model is given by Equation 1. [18]

$\sqrt{u}=\sqrt{\tau Y/\gamma}+\sqrt{K} \rightarrow (1)$

Where is the dynamic viscosity, γ the shear strain rate, yield stress and the viscosity consistency. The yield stress of human blood in normal condition is between 0.0003 Pascal and 0.02 Pascal. The yield stress was taken as 0.004 Pascal. The other properties and boundary conditions were assumed same as Newtonian model simulation condition. [19]

Conclusion

In this study, the CFD models of aorta are constructed to investigate the non-Newtonian nature of blood on a pulsatile flow. Results of velocity distribution, wall pressure and WSS distribution of aorta at four different time instants in the cardiac cycle have been presented for Newtonian and non-Newtonian fluid models. It is concluded that the flow patterns of Newtonian and non-Newtonian blood models are similar, but the non-Newtonian nature of blood caused a significant increase in wall Shear Stress (WSS) patterns. It is very difficult to observe the quantitative information of hemodynamic profiles like flow parameters, wall pressure and WSS in vivo. Computed profiles from the aorta model could be used as a diagnostic tool in clinical applications. For example, the measured flow profiles from a diseased subject could be compared with flow profiles from a healthy subject. The information generated from this comparative study plays an important role in understanding of the pathologic condition of diseased subject. The computed hemodynamic profiles from the CFD analysis could also be used with surgery and anesthesia simulators to train the medical professionals.

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