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Effect of six non-Newtonian viscosity models on hemodynamic parameters of pulsatile blood flow in stenosed artery

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Abstract
A numerical study of hemodynamic parameters of pulsatile blood flow is presented in a stenotic artery with non-Newtonian models using ADINA. Blood flow was considered laminar, and the arterial wall was considered rigid. Studied stenosis severities were 30, 50, and 70% of the cross-sectional area of the artery. Six non-Newtonian models were used to model the non-Newtonian behavior of blood, and their results were compared with the Newtonian model. The results showed that in Power-law and Walburn-Schneck models, unlike other models, shear stress values before and after the stenosis were smaller than Newtonian models. Also, in maximum flow rate, the Carreua, generalized Power-law, Casson, and Carreua-Yasuda models showed a reduction in global importance factor of non-Newtonian behavior, and subsequently, the results approached Newtonian model. In minimum flow rate, the global importance factor of Newtonian behavior increased, which highlighted the importance of Newtonian model. In minimum flow rate, Carreua-Yasuda model was more sensitive to the non-Newtonian behavior of blood compared to Carreua, Casson, and Power-law models. Also, in that time period, Walburn-Schneck was less sensitive to the non-Newtonian behavior of blood. On the other hand, this model did not show sensitivity when the flow rate was at its peak. Power-law model overestimated the global importance factor values. Therefore, Power-law model was not suitable, because it showed extreme sensitivity to dimension. Walburn-Schneck model was not suitable too because it lacked sensitivity.

1. Introduction

Blood is a heterogeneous multiphase mixture of solid blood cells (red blood cells, white blood cells, and platelets) in the fluid plasma, which has several non-Newtonian rheological properties such as the dependence of the deformation rate, viscoelasticity, and yield stress [1]. Blood viscosity depends on various factors such as the viscosity of plasma, hematocrit percentage, and distribution and mechanical properties of blood cells [2-4]. Non-Newtonian properties of blood

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affect the stress on artery walls, which can facilitate the onset and progression of lesions such as deposits inside the artery walls [5]. No model, whether Newtonian and non-Newtonian, can cover all complex properties of blood. Therefore, several different models are being used. These models have differences and can yield different results [6-8]. The capability of non-Newtonian models is different in covering physical phenomenon.

Different hypotheses have been proposed for the relationship between hemodynamic parameters and the emergence and development of the disease, which focus primarily on wall shear stress values [9]. Several numerical studies have also been conducted in this area. For example, Ishikawa et al. [10] numerically investigated the effect of non-Newtonian pulsatile blood flow through a stenotic vessel. They assumed a compressible and laminar flow. They investigated wall shear stress, flow pattern, the separation region, and pressure distribution, and stated that non-Newtonian properties of blood reduce the power of the vortex which develops downstream of the stenosis.

Mallik et al. [11] used the Power-law model to analytically solve the blood flow through a vessel. They investigated the parameters such as fluid velocity, volumetric flow rate, wall shear stress, and effective viscosity. Jung et al. [12] studied the non-Newtonian blood flow in a stenotic artery, and compared the results with an experimental study and obtained acceptable results. The effect of stenosis severity and wall shear stress was also examined. Sapna [13] analytically solved the non-Newtonian blood flow in a stenotic artery. Wall shear stress and apparent viscosity were investigated. It was found that increase in stenosis severity increases the apparent viscosity and wall shear stress, but due to the non-Newtonian properties of blood, this increase is not significant. Amornsamankul et al. [14] studied the non-Newtonian behavior of blood in a stenotic artery with a porous wall and investigated the significance of the assumption of non-Newtonian blood. The stenosis severity was 50% asymmetric and symmetric. They used Carreua model which is non-Newtonian. Pressure drop, axial velocity, and wall shear stress were investigated.

It is widely accepted that at shear rates higher than 100 s^{-1}, blood behaves as a Newtonian fluid [9, 15]. Although a study by Johnson et al. [16] showed that at low velocities of blood flow, non-Newtonian effects become important. Since there are periods during the cardiac cycle in which the flow is slow, it is reasonable that non-Newtonian effects of blood flow during this period become important. It is therefore essential to know the extent to which the non-Newtonian effects become important in the cardiac cycle. The purpose of this study was to investigate the effect of stenosis percentage on wall shear stress of an artery, global importance of non-Newtonian effects, pressure drop, and velocity profile using six non-Newtonian models.

2. Present work and numerical methods

The present study is investigated the pulsatile and laminar blood flow in a rigid artery with stenosis using Adina 8.8 finite element software. The general form of the governing equations of fluid motion (momentum equation and continuity equation) is as follows [17]:

$$\frac{\partial(\rho V)}{\partial t} + \nabla.(\rho V V - \bar{\tau}) = f_B \quad (1)$$

$$\frac{\partial \rho}{\partial t} + \nabla.(\rho V) = 0 \quad (2)$$

where, t is time, ρ is density, V is the velocity vector, f_B is vector of volumetric force exerted on the fluid (assumed to be zero here), and τ is the stress tensor which is calculated from the following equation:

$$\tau = -PI + 2\mu e \quad (3)$$

$$e = \frac{1}{2}(\nabla V + \nabla V^T) \quad (4)$$

where, P is pressure, I is the unit matrix, and μ is the apparent viscosity of the fluid (blood). Newtonian viscosity is equal to 0.0033 Pa.s. The non-Newtonian fluid models which were used to model the non-Newtonian behavior of blood are Power-law [18], Carreau [18], Carreau-Yasuda [19], Modified-Casson [20], Generalized Power-law [21], and Walburn-Schneck [22] that respectively are shown in the below equations:

$$\mu = k|\dot{\gamma}|^{n-1} \quad (5)$$

$$k = 0.035, n = 0.6$$

$$\mu = \mu_\infty + (\mu_0 - \mu_\infty)(1 + A|\dot{\gamma}|^2)^n \quad (6)$$

$$A = 10.976, n = -0.3216,$$

$$\mu_\infty = 0.0033, \mu_0 = 0.056$$

$$\mu = \mu_i + \frac{\mu_0 - \mu_i}{[1 + (\lambda|\dot{\gamma}|^b)]^a}$$

$$\mu_0 = 0.016, \mu_i = 0.0033, \lambda = 8.2, \quad (7)$$

$$b = 1.23, a = 0.64$$

$$\mu = \left[\sqrt{\tau_y\left(\frac{1 - e^{-m|\dot{\gamma}|}}{|\dot{\gamma}|}\right)} + \sqrt{\mu_c}\right]^2 \quad (8)$$

$$\mu_c = 0.0033, m = 100$$

$$\mu = k(\dot{\gamma})|\dot{\gamma}|^{n(\dot{\gamma})-1}$$

$$k(\dot{\gamma}) = \mu_\infty + \Delta\mu \exp\left[-\left(1 + \frac{|\dot{\gamma}|}{a}\right)\exp\left(\frac{-b}{\dot{\gamma}}\right)\right]$$

$$n(\dot{\gamma}) = n_\infty - \Delta n \exp\left[-(1 + \frac{|\dot{\gamma}|}{c})\exp(\frac{-d}{\dot{\gamma}})\right] \quad (9)$$

$$\mu_\infty = 0.0033, \Delta\mu = 0.25, n_\infty = 1, \Delta n = 0.45, a = 50, b = 3, c = 50, d = 4$$

$$\mu = a_1 \exp(a_2 + \frac{a_3}{H^2})\ |\dot{\gamma}|^{(1-a_4 H)}$$

$$a_1 = 0.00797\ \mathrm{Pa.s}, a_2 = 0.0608, a_3 = 377.7515, a_4 = 0.00499 \quad (10)$$

In this paper, a model of the right coronary artery with symmetric stenosis and rigid wall is considered. Computational domain is shown in Fig.1.

The severity of stenosis was 30, 50, and 70%. The equation which describes the shape of stenosis is [23-25]:

$$\frac{R(z)}{R_0} = 1 - (\frac{R_0 - R_{0,t}}{2R_0})(1 + \cos\frac{2\pi(z - z_m)}{L_{st}}) \quad (11)$$

where, R_0 is the radius healthy vessel and is equal to 0.015m, R(z) is the radius of the vessel in stenotic region, $R_{0,t}$ is the radius of the vessel at the beginning of the stenosis, z_m is the axial coordinates relative to the origin of coordinates, and L_{st} is length of stenotic region. Also, the length of entrance region was $20R_0$, the length of the stenotic region was $4R_0$, and the length of the entrance region was $42.66R_0$. Blood was considered as incompressible fluids with a density of 1050 kg/m^3.

Figure 2 shows the average velocity profiles applied on the inlet boundary condition [23-25]. Time, in Fig. 2, has become dimensionless with heart fluctuation period which is 0.8 s.

Grid independence of results was examined to obtain grid independent results to determine the best results and to minimize computer running-time. Three inlet velocity pulses were considered to achieve stable and convergent solutions [26]. The simulations were performed on a computer equipped with a 2.93 GHz processor and 8 G of RAM.

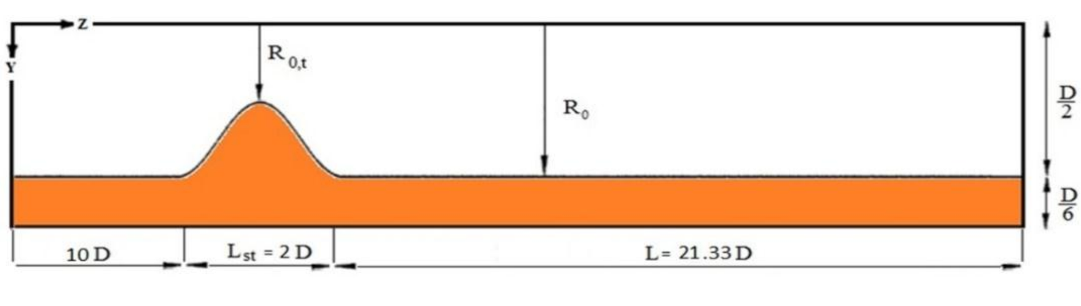

Fig. 1. Computational domain for an artery with stenosis and its dimensions.

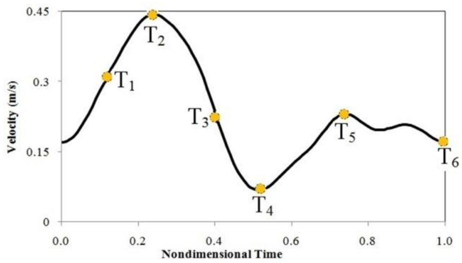

Fig. 2. Inlet velocity pulse [23-25].

3. Validity of the solution

A numerical study by Jeong et al. [27] was used to check the validity of numerical methods used in the stenotic unsteady case. They investigated the left coronary artery. In their study, blood vessel wall was assumed to be rigid, and blood was assumed to be Newtonian fluid. The comparison of dimensionless pressure drop and shear stress between the results of this study and that of Jeong et al. [27] are shown in Fig. 3. The parameters were made dimensionless using dynamic pressure, which was calculated based on the average velocity. As can be seen, the results of this study and that of Jeong et al. [27] are in a good agreement.

4. Results and discussion

4.1. Important global factor

IG, or global non-Newtonian importance factor, is defined to allow for quantitative comparison between different non-Newtonian and Newtonian models [28-30]:

$$\mathrm{IG} = \frac{1}{N}\frac{\left[\sum_{N}(\mu-\mu_\infty)^2\right]^{\frac{1}{2}}}{\mu_\infty} \tag{12}$$

In the above equation, all N nodes which were encompassed in the region are used to determine the importance factor of non-Newtonian behavior. In that equation, μ is the effective viscosity, and μ_∞ is Newtonian viscosity, which is equal 0.0035 in the present study. It should be noted that in Johnston et al. [29], N was the number of nodes on the vessel wall. On the other hand, in this numerical study, according to Razavi et al. [28] and Soulis et al. [30], N is the total number of points in the computational domain, so that the effects of different number of nodes on the wall would be reduced in comparison between different cases. It is important to find a criterion, above which the assumption of non-Newtonian fluid can be made.

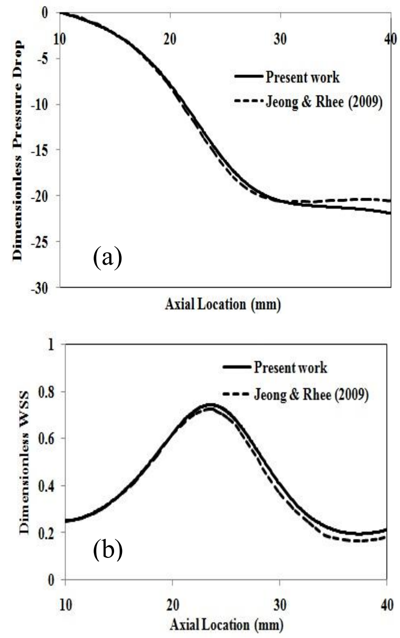

Fig. 3. Axial dimensionless; (a) Pressure drop and (b) Wall shear stress.

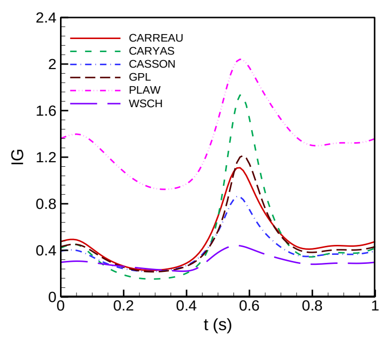

Fig. 4. IG for 30% stenosis.

4. 2. Effect of stenosis percentage on IG

Figure 4 shows the importance factor of non-Newtonian behavior in terms of time for different models in a 30% stenosis severity. Power-law model yielded higher values compared to other models, which is associated with the trend of viscosity variations in terms of shear rate. That is because even at high shear rates, there was a difference between the Power-law model and the Newtonian model, but in models like Carreua-Yasuda and Casson, viscosity approached to

Newtonian viscosity with increasing shear rate. Walburn-Schneck model yielded lower values compared to other models. It was also associated with the trend of viscosity in this model because the viscosity values obtained from this model were lower compared with other models.

As shown in Fig. 4, in peak flow rate, the Carreua, generalized Power-law, Casson and Carreua-Yasuda models showed a reduction in IG, and thus, they approached the Newtonian model. In minimum flow rates, these models showed an increase in IG, and thus, showed the importance of Newtonian behavior. In minimum flow rate, Carreua-Yasuda model was more sensitive to the non-Newtonian behavior of blood compared to Carreua, Casson, and Power-law models. In this period, Walburn-Schneck was less sensitive to the non-Newtonian behavior of blood. On the other hand, in the time period of maximum flow rate, increase in IG, and consequently, diverging from the Newtonian model was seen in this model. Therefore, Power-law model is not suitable, because it shows extreme sensitivity to dimension. Also, Walburn-Schneck model is not suitable, because it lacks sensitivity.

Table 1 shows the values of global importance for the six non-Newtonian models. The comparison of the values of global importance in different stenosis severities shows that IG does not follow the same trend in different models. For example, in Casson and Carreua-Yasuda models, this factor increases with increase in stenosis severity. In Power-law and Walburn-Schneck models, this factor decrease with increase in stenosis severity, which is associated with changes in the viscosity in terms of shear rate for each of these models. IG is a criterion, i.e. below IG, the fluid behavior is Newtonian, and above that, the behavior is non-Newtonian. According to Table 1, IG was calculated as 0.4. In numerical studies conducted by Razavi et al. [28], Johnson et al. [29], and Soulis et al. [30], IG was determined as 1, 0.25, and 0.25, respectively.

4.3. Effect of non-Newtonian model on axial velocity profile

The results discussed in this section were obtained from the model with 70% stenosis severity. In this model, a major part of the luminal area was affected by separation phenomenon. The stenosis makes it difficult for flow to pass the vessel and can cause problems in medical context. Figure 5 shows the axial velocity profiles at sections 1 through 4 for the Newtonian model, as well as six non-Newtonian models, at t=0.24s. Section 1 is the throat of stenosis, section 2 is at one diameter distance from the neck, section 3 is at two diameter distance from the throat, and section 4 is at four diameter distance from the throat. The comparison between the graphs shows that there is a slight difference between Newtonian model and four non-Newtonian models: Carreua, Carreua-Yasuda, Casson, and generalized Power-law. Power-law model showed the lowest axial velocity.

4.4. Effect of stenosis percentage on pressure drop

The trend of variations in arterial pressure in the axial direction is compared in Fig.6 between Newtonian and six non-Newtonian models with stenosis percentages of 30, 50, and 70%. According to Fig. 6, with an increase in stenosis severity, pressure increases in the first half of the stenosis. There is also a sharp drop in pressure due to the stenosis. If the percentage of arterial stenosis and pressure gradient along the artery is high enough, the pressure may be negative at the neck. Negative pressure can damage the plaques. As can be seen, Power-law and Walburn-Schneck models showed lower pressure difference (inlet-outlet) compared with other models. These differences can affect the stresses and strains in these regions.

4.5. Effect of stenosis on wall shear stress

Time-averaged shear stress (Mean WSS) is a hemodynamic parameter that has been used in different studies to determine atherosclerosis-prone areas. It is defined as follows [28]:

$$\text{Mean WSS} = \frac{1}{T}\int_0^T \tau_w \, dt \qquad (13)$$

where, T is the period of cardiac cycles, and τ is the shear stress vector. Regions with low time-averaged shear stress (less than 1 Pa), or with high shear index are known to be prone to atherosclerosis [31]. Figure 7 shows the graphs of time-averaged shear stress and oscillatory shear index for the six non-Newtonian models compared to Newtonian model for 50% stenosis severity. Increase in time-averaged shear stress can increase nitric oxide production by endothelial cells [32]. Nitric oxide protects endothelial cells from apoptosis and leads to the proliferation of these cells to cover plaque's surface during plaque growth. As can be seen from Fig. 7, at downstream and upstream of the stenosis, Carreua, Carreua-Yasuda, and Casson models estimated greater shear stress values compared to Newtonian Model. At the stenosis region, where shear stress increases significantly, these models showed values similar to the Newtonian model, and their sensitivity to the non-Newtonian behavior of blood was good.

Generalized Power-law model showed values closer to Newtonian model, but, before and after the stenosis, the Power-law and Walburn-Schneck models showed lower shear stress values compared to the Newtonian model. Also in the stenosis region, these models showed lower shear stress values compared to the Newtonian model, which is not correct.

As can be seen, Carreua and Casson models showed almost similar results. Generalized Power-law model showed variations close to the Newtonian model. The small values of time-averaged shear stress (less than 1 pa) and high oscillatory shear stress increase rupture in intercellular links, as well as permeability in the cells in these regions [33]. Low time-averaged shear stress values show atherogenic conditions in post-stenosis region.

Table 1. IG for different stenosis percentage.

% Stenosis	Walburn–Schneck	Power-law	Generalized Power-law	Casson	Carreau–Yasuda	Carreau
30	0.300	1.334	0.459	0.395	0.486	0.477
50	0.294	1.268	0.457	0.404	0.488	0.465
70	0.286	1.181	0.454	0.421	0.496	0.448

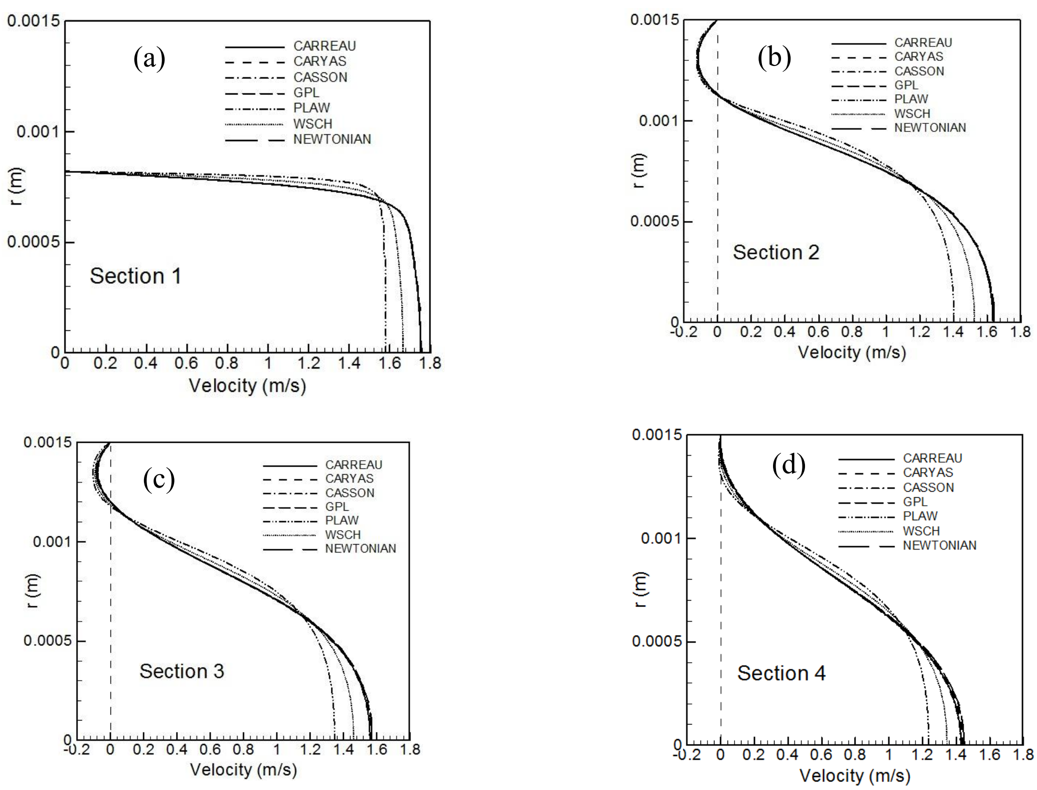

Fig. 5. Axial velocity profile, t=0.24s; (a) section 1, (b) section 2, (c) section 3, and (d) section 4.

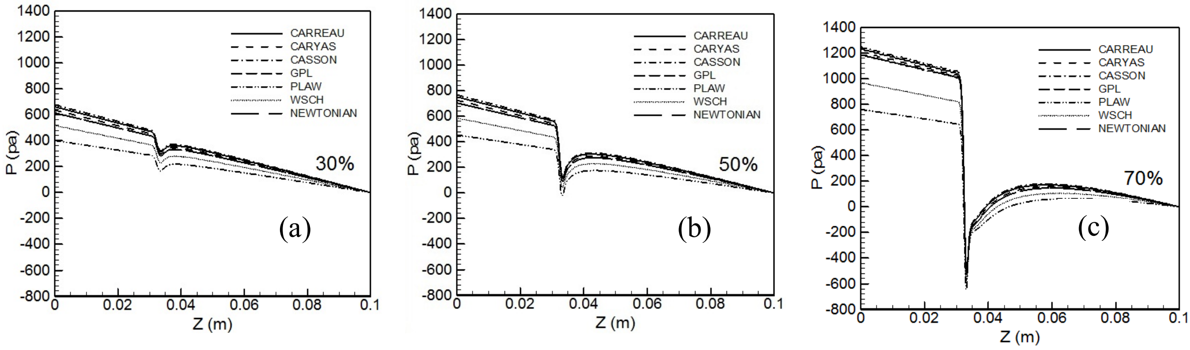

Fig. 6. Effect of stenosis percentage on pressure drop, t=0.24s; (a) 30%, (b) 50%, and (c) 70%.

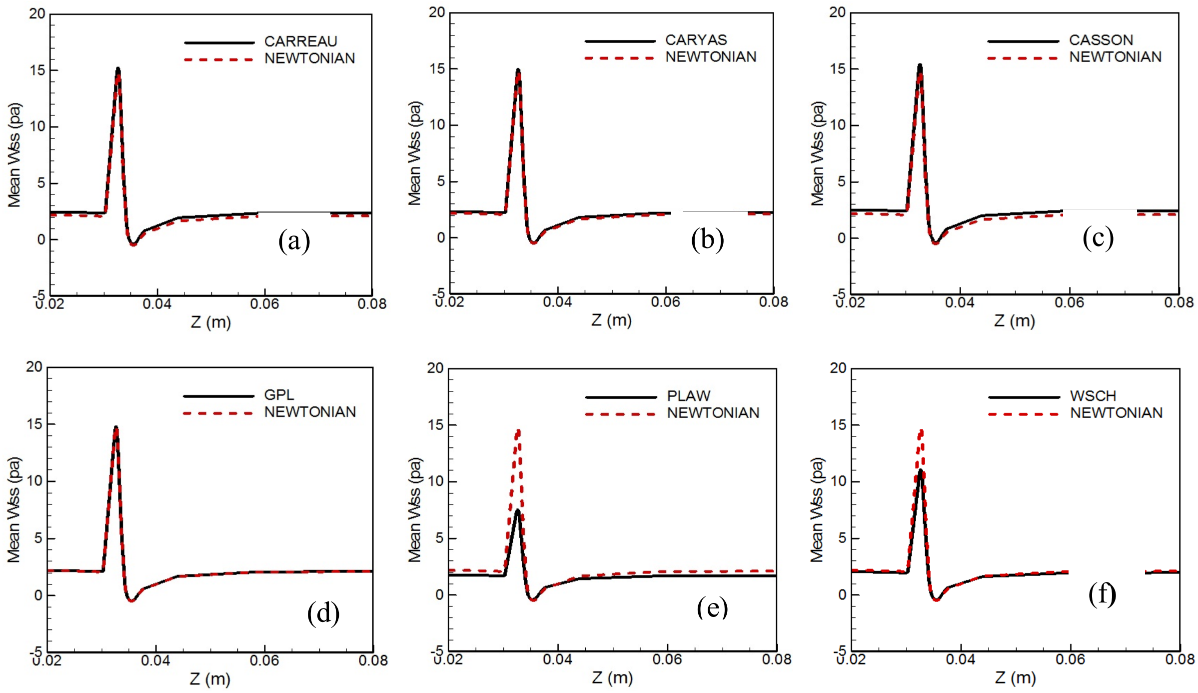

Fig. 7. Mean WSS, 50% stenosis; (a) Carreau, (b) Carreau-Yasuda, (c) Casson, (d) Generalized Power-law,(e) Power-law, and (f) Walburn-Schneck.

5. Conclusions

In this paper, ADINA 8.8 software was used to investigate pulsatile non-Newtonian blood flow in arteries with symmetrical 30, 50, and 70% stenosis. The results showed that Power-law model yielded higher values of IG compared to other models, which is associated with the trend of viscosity variations in terms of shear rate. Also, the comparison of the values of IG in different stenosis severities showed that IG does not follow the same trend in different.According to the results, IG criterion was selected to be 0.4. The results showed that the Power-law and Walburn-Schneck models have the lowest axial velocity, the lowest pressure difference between inlet and outlet, and the lowest shear stress.

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