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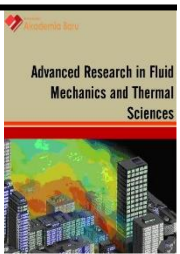

Simulation of non-Uniform Magnetic Field Impact on non-Newtonian and Pulsatile Blood Flow

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ABSTRACT

Blood contains haemoglobin. Haemoglobin is an iron (Fe) containing protein that respond to magnetic field. Regarding imposing blood flow upon magnetic field is not obvious well and or has not been examined completely and extensively and concerning contradictory results that presented by researchers, in this paper, laminar, pulsatile and non-Newtonian blood flow imposed by non-uniform magnetic field were explored with different intensities through using COMSOL 4.4 Finite Element Software. Non-Newtonian Generalized power-law model was applied and with aim of comparison, results were compared with Newtonian mode. Findings indicated that wall shear stress fluctuation amplitude was increased by raising magnetic field intensity for both Newtonian and non-Newtonian modes and reverse flow occurred for 4 and 6 T field's intensity. Concerning non-Newtonian mode, as the intensity of the magnetic field increases from 0.5 to 1, 1 to 2, 2 to 4, and 4 to 6, the amplitude of shear stress fluctuations will rise to 1.136 , 1.729, 3.969, and 1.626 times respectively since non-Newtonian fluid viscos dissipation is higher. Moreover, increased average arterial pressure was achieved in non-Newtonian mode more than Newtonian mode.

1. Introduction

Todays, thickening intima layer and narrowing of artery is the main reason of cardiovascular diseases and formation of Atherosclerosis and creation of disorder in normal blood flow and consequently people death due to macromolecule penetration into artery wall and formation of fat mass. There are various ways for controlling and adjusting blood flow in this case [1] that dissolving fat mass and increasing heat transfer rate are the most common ways [2]. Other solution is to remove created vorticities after fat mass that interfere with main blood flow [3] that this operation is performed by an external magnetic field and applying extra force to biological fluids such as synovial fluid [4] and spinal fluid [5]. During past decades, numerous studies have been conducted in this regard that in most cases, non-Newtonian effects of bio-fluid and porosity of vascular wall had

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neglected [6-8]. In Tzirtzilakis and Loukopoulos work [9], impact of applied external magnetic fields was explored on speed and temperature of bio-fluid in artery. Recently, scholars followed non-Newtonian models for bio-fluids to simulate biological phenomena well [10-17]. Accordingly, different methods including viscoelastic fluids, power law and etc. for simulation. In this regard, it can refer to Mirsa work [16] that was conducted concerning application of viscoelastic model in analyzing bio-fluids behaviors affected by magnetic field.

Shit *et al.,* [18] examined analytically effect of magnetic field on incompressible micropolar fluid with low Reynolds in an asymmetric channel. Exact analytic solutions were obtained for axial velocity parameters, streamlines patterns, magnetic field function and inductive magnetic field. Results indicated that with increasing magnetic field intensity, the decreased speed and the trapped fluid was eliminated in the rotational flow.

Alimohammadi *et al.,* [5] explored impact of magnetic field on 2-D and steady flow of a bio-magnetic non-Newtonian fluid into stenosed porous artery. Power law model was employed for simulating non-Newtonian fluid. Stenosis caused a reverse flow in back area of stenosis and disturbance into artery flow that external magnetic field employed for dampening of these vorticities and uniformity of flow. Moreover, shear stress imposed on artery wall was decreased due to the neutralization of generated vorticities caused by the production of vorticities in the opposite direction to the existing vorticities by the applied magnetic field.

Sharma *et al.,* [19] presented a mathematical model for problem of targeting magnetic drug in a channel. Nanoparticles were injected to channel and affected by magnetic field. Furthermore, they used COMSOL Finite Element Software to analyze direction and impact of nanoparticles drug carrier affected by magnetic field. Simulation results like findings of mathematical model was indicator of the efficiency improvement of the particle placement at the desired location by increasing the intensity of the applied magnetic field.

Hasanzade [20] studied the non-Newtonian flow of blood influenced by incompressible magnetic field in a two-dimensional artery, aiming to attach the drug to magnetic particles and direct them through the magnetic field to the desired point. He assumed the blood as Viscoplastic. The simulation results showed that due to the rotational flow formed near the magnetic source, the absorption of drug in the target area was increased. By comparing the simulation results for the Casson and Newtonian models, it was observed that in non-Newtonian model, the formed rotational flow is smaller than those in Newtonian model, the Newtonian model speed is about 5.7% higher than the non-Newtonian model, the shear stress of the Newtonian model is less than the non-Newtonian model, and the shear force of the non-Newtonian model is about 36% higher than the Newtonian model.

Lunnoo and Puangmali [21] investigated the computer simulation of magnetic drug targeting using the COMSOL 4.4 Finite Element Software. The parameters studied were the size of the magnetic nanoparticles, the type of magnetic nanoparticles and the coating thickness attached to the nanoparticles. Blood flow was assumed to be non-Newtonian by the generalized Power law model and examined a two-dimensional model of a rigid blood artery with different branches. They investigated the pulsatile and laminar flow under a magnetic field of a magnet and used 5 speed pulses to achieve stable oscillation solution. The results showed that the absorption efficiency of nanoparticles decreased with decreasing their size. Also, since it is very difficult to work with nanoparticles with a diameter of less than 20 nm, it is suggested that this method be used for small arteries and low blood velocity.

Park [22] with aiming to transfer the drug to the stenosed area, using the Ansys-CFX software, analyze the fluid-solid interaction in the right curved coronary artery. A three-dimensional model of a curved stenosed artery was reconstructed based on clinical angiography images. For a better comparison, the artery also assumed rigid and used from four non-Newtonian models of Power law, Carreau, Casson and Walburn-Schneck. Maximum of total displacement and von Mises stress were calculated 2.14 mm and 92.06 Mpa, respectively. In addition, no significant difference was observed between the sedimentation of the particles in the stenosed area both in rigid and elastic modes.

In recent years, many researchers have studied the non-Newtonian behavior of blood flow in coronary and carotid arteries. For example, Shirmer *et al.,* [23] conducted a study in a stenosed carotid artery by using Carreau model and Jeong *et al.,* [24] in the healthy coronary artery applying Carreau model. Furthermore, Guerciotti and Vergara [25] compared the calcification of Newtonian and non-Newtonian laminar blood flow in 3-D and rigid stenosed arteries. Their examined geometries were a carotid artery with severe stenosis and a coronary artery with different peripheral stenosis treated by bypass. They used the Caro-Yasuda model to simulate non-Newtonian blood behavior. The results showed that there is little difference in carotid artery geometry between the assumption of Newtonian and non-Newtonian blood mode (30% in vorticity field and 24% in the velocity field). While this difference was higher in the coronary artery.

Considering the importance of the mentioned cases and the fact that so far no studies have been done on the effect of the non-uniform magnetic field on the blood as a non-Newtonian fluid, this paper examined the interaction of the non-uniform external magnetic field with the blood flow containing magnetic carrier materials. Blood is considered as a continuous environment and the problem is addressed by solving the Maxwell and Navier-Stokes equations. The solution is that in the first stage Maxwell's equations are solved for the entire range of solution (permanent magnetism, blood arteries, tissue, and air), and in the second stage, a magnetic volumetric force that is due to the effect of the magnetic field on the fluid, is added to the Navier-stokes equations.

2. Definition of the Problem

The schematic of simulated geometry which includes the blood artery, the permanent magnet, the tissue around the artery and the air, is shown in Figure 1 in a 2-D View. Direction of blood entrance into artery is from left.

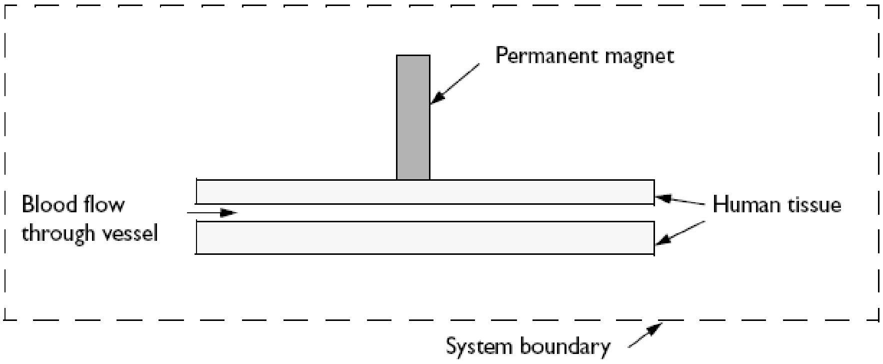

Fig. 1. 2-D Schematic of Simulated Geometry

The pulse of velocity input to the solution domain [26,27] is shown in Figure 2. As for the solution of fluid flow by finite element softwares, it is recommended that at the time between zero to one second, a linear graph be used for velocity in which it starts from speed of zero at time zero reaching the inception of physiologic pulse at time one. This will help to expedite convergence of solution or can even prevent solution divergence so that the fluid gradually enters the solution filed and reaches its real value in the beginning of physiological pulse. COMSOL Software calculates the magnetic field created by the permanent magnet and effects the magnetic volumetric force produced by it in the blood flow. Table 1 shows the properties of materials used for modeling.

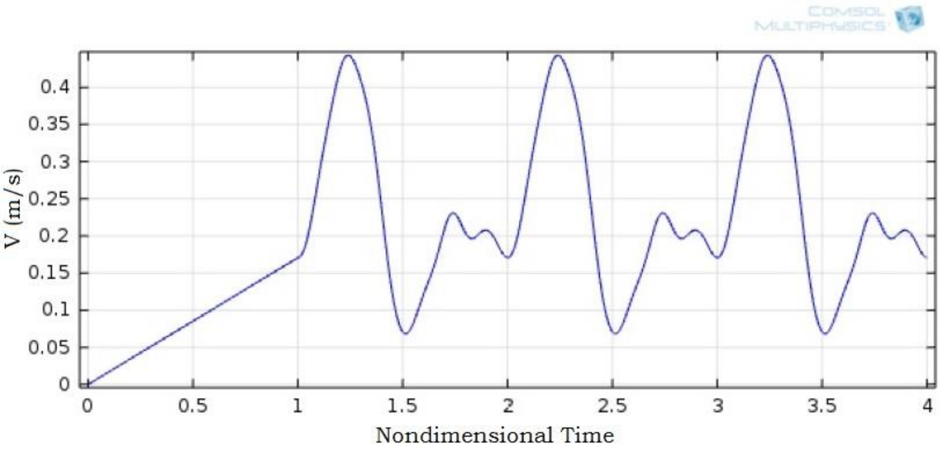

Fig. 2. Input velocity pulse

Table 1
Modeling Data

Quantity	Description	Value
$\mu_{r,mag}$	Relative permeability, magnet	5×10^{3}
B_{rem}	Remanent flux density, magnet	0.5T
α	Ferrofluid magnetization-curve parameter	10^{-4} A/m
β	Ferrofluid magnetization-curve parameter	3×10^{-5} $(A/m)^{-1}$
ρ	Density, blood	1060 kg/m^{3}
μ	Dynamic viscosity, blood	3.5×10^{-3} kg/(m.s)

In the present work, the problem was first simulated in the absence of a magnetic field for two Newtonian and non-Newtonian modes, and then for analyzing effect of the magnetic field intensity, the problem was simulated for 0.5, 1, 2, 4 and 6 T modes. The intensity of 6 Tesla is created by new MRI scanner machines. A person's exposure to such machines for a few minutes does require examining the intensity of Tesla field. It was already stated that exposure to such a field can be harmful; but the degree of damage or negative impact of such a field on hemodynamic parameters had not been examined. Magnetic fields less than 6 Tesla result from exposure to high voltage electricity cables, etc. A magnetic field more than 6 Tesla cannot be created easily and people are not usually exposed to them. For instance, a field of 100 Tesla belongs to a star called White Dwarf.

Dimensions and sizes of the solution domain and the pattern of meshing are shown in Figures 3 and 4. As shown in Figure 4, an unstructured mesh has been used and the grid has been thinner to increase the accuracy of the solution where more accurate computations are required.

3. Governing Equations

3.1. Equations Governing Magnetism

Given that the magnetic section is static, the Maxwell-Ampere's law for the magnetic field H (ampere per meter) and the current density J (amperes per square meter) is used [28].

$$\nabla \times H = j \tag{1}$$

In addition, the Gauss' law for the magnetic flux density B (Vs/m^{2}) states that

$$\nabla \cdot B = 0 \tag{2}$$

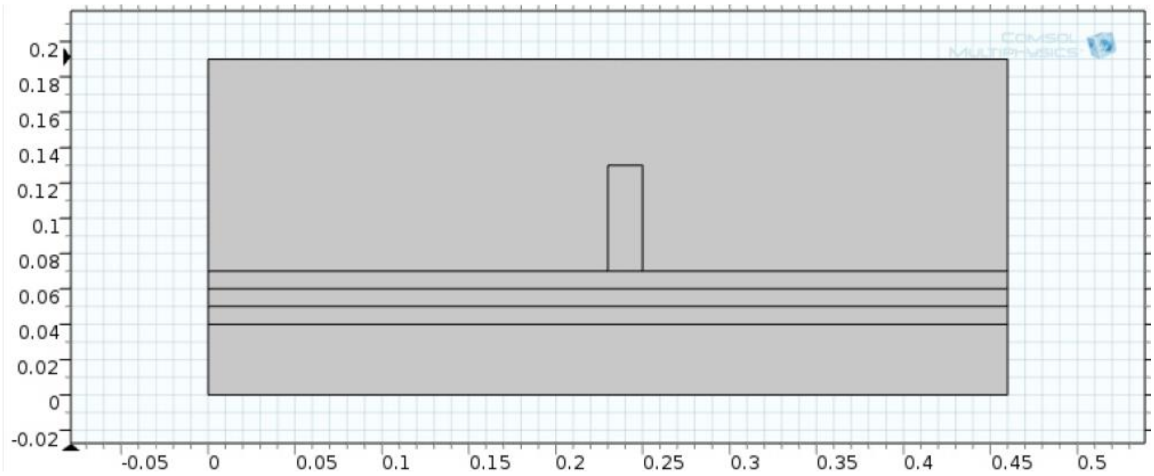

Fig. 3. Dimensions and Sizes of Solution Domain

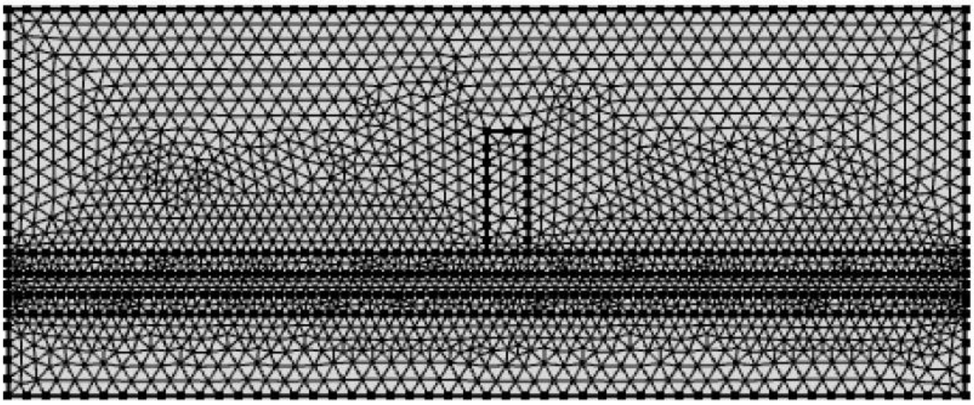

Fig. 4. Solution domain meshing

Constitutive equations describe the relationship between B and H in different parts of the solution domain as follows

$$B = \begin{cases} \mu_0 \mu_{r,mag} H + B_{rem} & \text{For permanent magnet} \\ \mu_0 \left(H + M_{ff}(H) \right) & \text{For blood flow} \\ \mu_0 H, & \text{For tissue and air} \end{cases} \tag{3}$$

In the above relation, μ_0 is the magnetic permeability coefficient of the vacuum based on Vs/m^2, μ_r the magnetic permeability of the magnetic magnet that is dimensionless, B_{rem} is the remanent magnetic flux based on A/M and M_{ff} is the magnetization vector in the blood flow based on A/M which is a function of the magnetic field H.

We define the potential of the magnetic vector (A) as in equation 4

$$B = \nabla \times A \ , \ \nabla \cdot A = 0 \tag{4}$$

Now we combine relations (1) and (3) together

$$\nabla \times \left(\frac{1}{\mu} \nabla \times A - M\right) = j \tag{5}$$

By simplifying for 2-D problems without any perpendicular flow

$$\nabla \times \left(\frac{1}{\mu_0} \nabla \times A - M\right) = 0 \tag{6}$$

It should be noted that the above equation assumes that the potential of a magnetic vector is only one non-zero member that is perpendicular to the plane, that is, A = A (0,0, A_z). If α based on $\frac{A}{m}$ and β based on $\frac{m}{A}$ in terms of two ferromagnetic fluid parameters and the induction of magnetization is also in the form of M (x, y) = (M_x, M_y) [29]

$$M_x = \alpha \ \arctan\left(\frac{\beta}{\mu_0} \frac{\partial Az}{\partial y}\right)$$
$$M_y = \alpha \ \arctan(\frac{\beta}{\mu_0} \frac{\partial Az}{\partial x}) \tag{7}$$

With linearization for the magnetic field, we study

$$M_x = \left(\frac{\alpha\beta}{\mu_0} \frac{\partial Az}{\partial y}\right)$$
$$M_y = -\left(\frac{\alpha\beta}{\mu_0} \frac{\partial Az}{\partial x}\right) \tag{8}$$

For the boundaries of the system that are far enough away from the magnet (shown in Figure 1 with the dash line), the magnetic field boundary condition ($A_z = 0$) can be applied.

3.2. Equations Governing the Fluid Part

Navier-Stokes equations for an incompressible fluid as follows

$$\rho \frac{\partial u}{\partial t} - \nabla . \eta (\nabla u + (\nabla u)^t) + \rho u . \nabla u + \nabla P = F$$
$$\nabla \cdot u = 0 \tag{9}$$

In the above equation, η is the dynamic viscosity based on kg/m.s, u velocity based on m/s, ρ is the density of the fluid in terms of kg/m^3, P is the pressure in terms of N/m^2, and F the volumetric force in terms of N/m^3.

With the assumption that there are no interactions between the magnetic nanoparticles in the blood, the force F = (F_x, F_y) for the weak fields is expressed by the relation $F = |m|\nabla|H|$. By combining equations (3), (4) and (8) we have

$$F_x = \frac{\alpha\beta}{\mu_0 \mu_r^2}\left(\frac{\partial A_z}{\partial x} \cdot \frac{\partial^2 A_z}{\partial x^2} + \frac{\partial A_z}{\partial y} \cdot \frac{\partial^2 A_z}{\partial x \partial y}\right)$$
$$F_y = \frac{\alpha\beta}{\mu_0 \mu_r^2}\left(\frac{\partial A_z}{\partial x} \cdot \frac{\partial^2 A_z}{\partial x \partial y} + \frac{\partial A_z}{\partial y} \cdot \frac{\partial^2 A_z}{\partial y^2}\right) \tag{10}$$

To calculate the volumetric force of blood flow, the two above sentences in the mass fraction of the K_{ff} ferrofluid should be multiply.

Since non-Newtonian blood flow properties are important for low blood flow velocities, or there may be times in a cardiac cycle that velocity is near zero, a suitable model for viscosity changes should be selected.

The generalized power law model calculates the shear wall stress for velocities of lower input and lower shear regions better than the Newtonian model which for the low to moderate shear range is very close to the Carreau model and acts as Newtonian for the medium to high shear range. This model defines the viscosity as follows [30-35]

$$\mu = 0.1\,\lambda|\dot{\gamma}|^{n-1} \tag{11}$$

where $\dot{\Upsilon}$ is the shear rate and n and λ are defined as follows

$$\begin{aligned}\lambda(\dot{\gamma}) &= \mu_\infty + \Delta\mu \exp\left[-(1+\tfrac{|\dot{\gamma}|}{a})\exp(-\tfrac{b}{|\dot{\gamma}|})\right]\\ n(\dot{\gamma}) &= n_\infty + \Delta n\, exp\left[-(1+\tfrac{|\dot{\gamma}|}{c})\, exp(-\tfrac{d}{|\dot{\gamma}|})\right]\end{aligned} \tag{12}$$

On the walls, the non-slip boundary condition (u = v = 0) and the boundary condition of the output are applied as P = 0. As previously explained, the input boundary condition is a velocity profile that to achieve a stable oscillation solution, 3 pulses are used [36, 37].

4. Results

Figure 5 shows the details of the magnetic field intensity. As it is shown, the greatest magnetic field intensity occurred inside the magnet. Figure 6 shows the profile of the blood flow velocity at the moment of 4s. As it is shown, the magnetic field causes deviation of blood flow.

Sometimes, stroke is caused by high blood pressure which results in the rupture of arteries that puts pressure on the local cerebral tissues as a result of bleeding. The increased thickness of the artery wall is a complication that is attributed to high blood pressure. Moreover, it is claimed that the tension of the artery in response to the high blood pressure can lead to atheroma and that the damaging effect of high blood pressure is more than low shear stress [35]. As seen in Figures 7 and 8 for the Newtonian and non-Newtonian modes, the magnetic field caused an increase in blood pressure in the area in which the magnet is located that this increase in pressure is greater with increasing magnetic field intensity. The result of the present work is in accordance with the work of Gmitrov [38], which after studying the effect of the 0.35 T field on the receptors of the rabbit's aorta arch, stated that the use of the magnetic field increased the average arterial pressure. Of course, as equation. 9 also states, this increase in pressure is obvious because of the force of F in the opposite direction of blood flow which reduces the velocity of blood flow and, consequently, increases arterial pressure. By moving away from the magnet and reducing the intensity of the magnetic field, the volume force on the blood flow reduced and the arterial pressure returns to normal mode.

Of course, it should be noted that the results obtained in this paper contradict the work of Xu *et al.,* [39]. After examining the amount of the blood velocity in the Tibialis anterior muscle of a mouse, Xu *et al.,* stated that the magnetic field of 1 mT increased the rate of blood flow by 20%-45%.

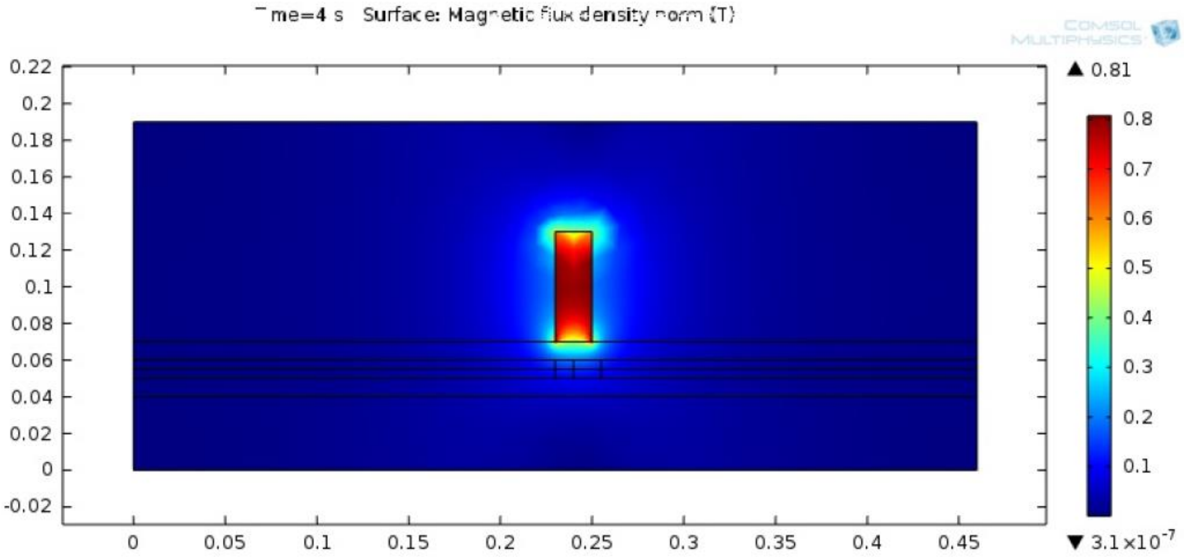

Fig. 5. Details of the intensity of the magnetic field at the moment 4s

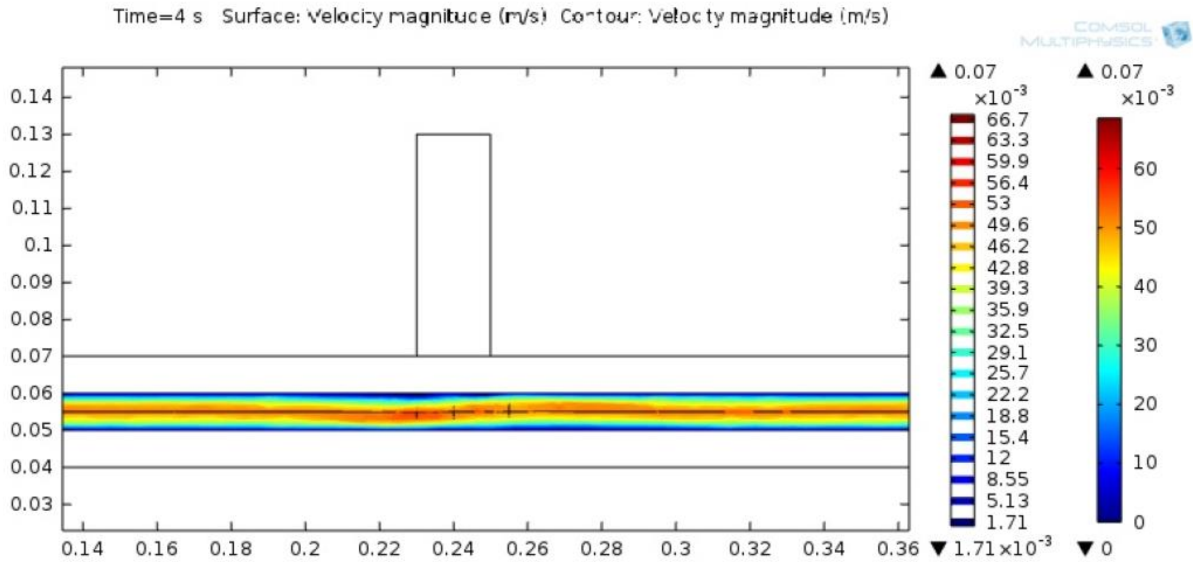

Fig. 6. Blood flow velocity profile at 4s moment

It is also seen by comparing Figures 7 and 8 that the pressure increase in non-Newtonian mode is greater than the Newtonian mode. The reason for this is that the studied non-Newtonian fluid is blood with a non-Newtonian generalized Power law model that its viscosity is slightly more or equal to the Newtonian model. Therefore, the viscous dissipation is the larger non-Newtonian fluid results in a higher pressure drop. The recent result is in agreement with Al-Habahbeh's work [40].

Considering its effect on the performance of the endothelial cells the inner artery walls, the stress applied to the wall of artery is one of the determinative hemodynamic factors in the progress of atherosclerosis [35]. The effect of the magnetic field on the wall shear stress (WSS) of the artery is shown in Figures 9 and 10 for non-Newtonian and Newtonian modes. For non-Newtonian mode, as shown in Figure 9, there is no negative WSS in the field intensity up to 2 T, and WSS fluctuations increase with increasing magnetic field intensity. With increasing intensity of the magnetic field from 2 T to 4 T and 6 T, the WSS oscillations become more intense and in the areas before and shortly after the placement of the magnet, the return flow region is formed and the amount of stress is negative.

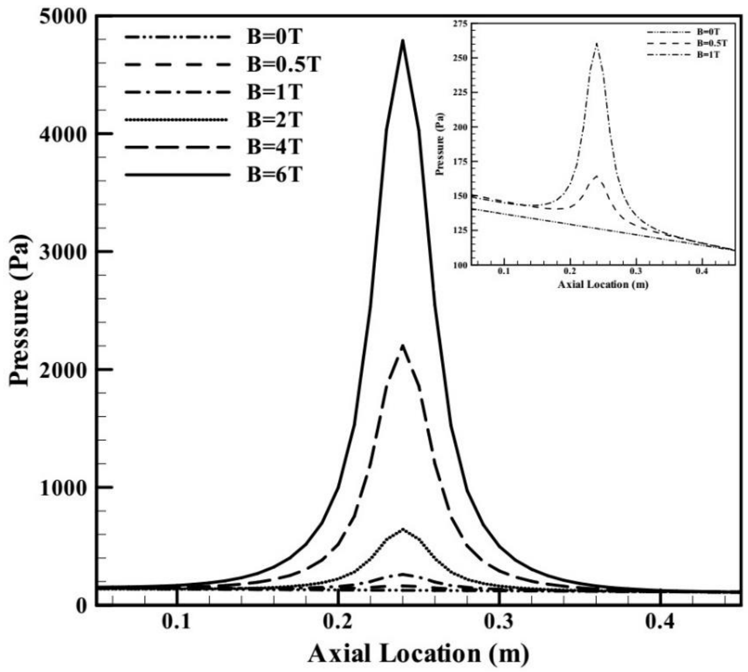

Fig. 7. Axial pressure drop of blood flow (Newtonian mode)

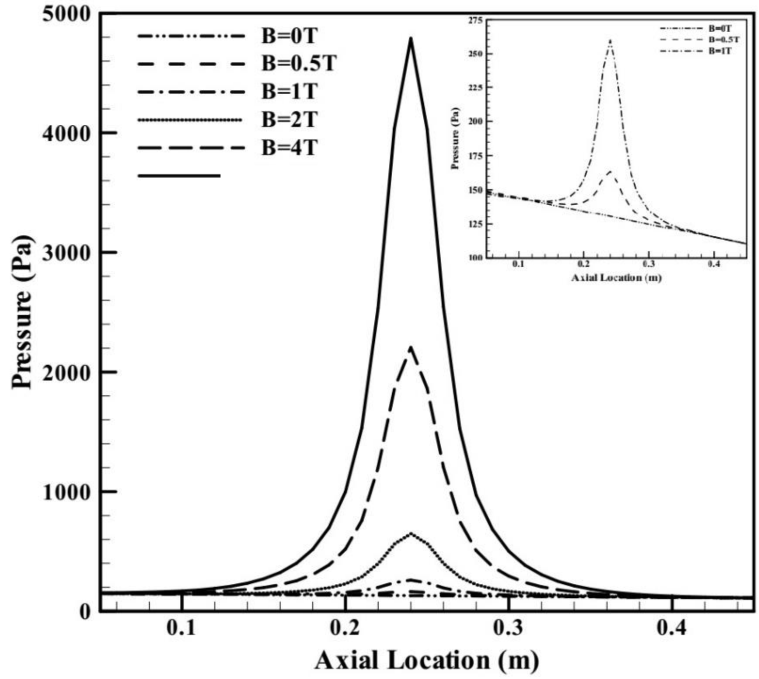

Fig. 8. Axial pressure drop of blood flow (non-Newtonian mode)

Table 2 shows the maximum and minimum WSS for non-Newtonian and Newtonian modes. As seen from non-Newtonian flow results, the 0.5 T magnetic field has a negligible effect on the flow of blood. By doubling the magnitude of the magnetic field, the WSS fluctuation range is 1.133. By increasing the intensity of the magnetic field and bringing it to 2 T, the effect of the field on the flow of blood becomes more apparent and the magnitude of the oscillation is equal to 1.732 relative to the 1 T mode. By increasing the intensity of the magnetic field and bringing it to 4 Ta, the effect of the field on the blood flow is intensified and the magnitude of the oscillation is equal to 3.965times compared to the 2 T.

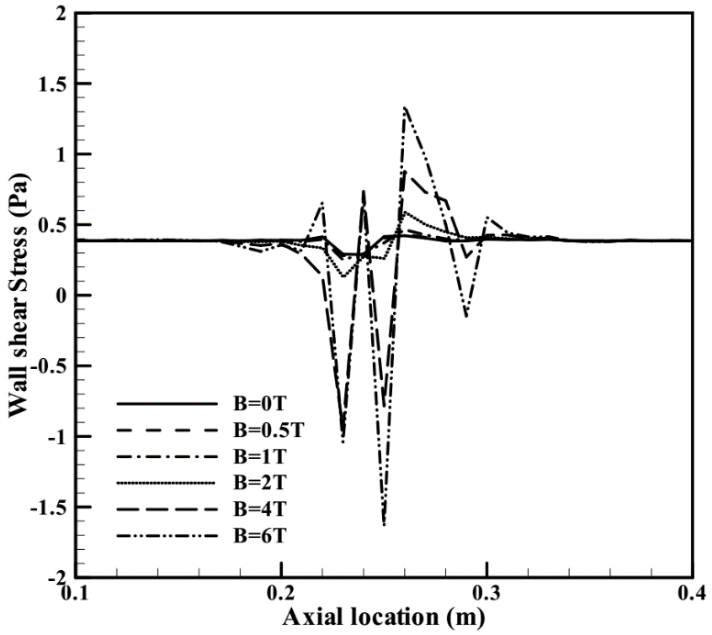

Fig. 9. WSS of artery (non-Newtonian mode)

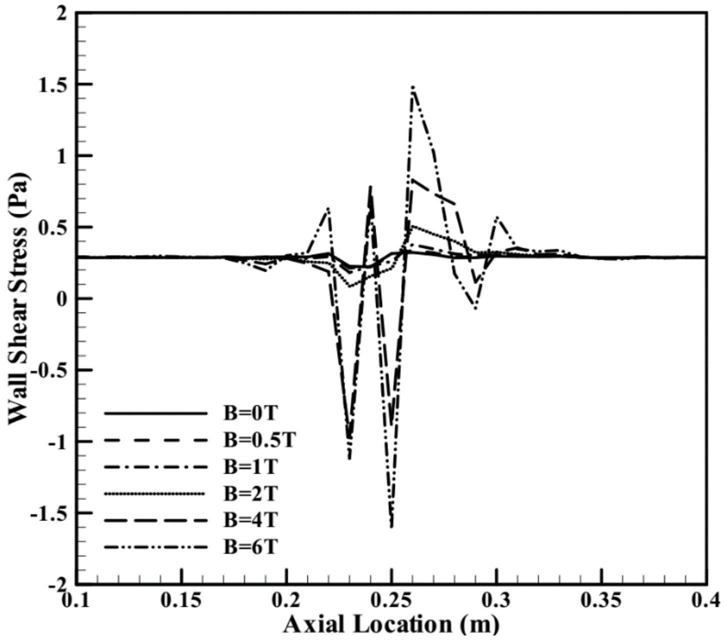

Fig. 10. WSS of artery (Newtonian mode)

Due to the large similarity of the results of the Newtonian mode with non-Newtonian mode, the description is ignored in this regard. Details of the Newtonian mode results are given in Table 2. According to Table 2 and the comparison of the Newtonian mode with non-Newtonian mode, it is observed that, up to the intensity of the 4 T field, the WSS in non-Newtonian mode is greater than the Newtonian mode, which according to Figure 11 and this fact that non-Newtonian viscosity is greater than Newtonian viscosity, this issue is justifiable.

Table 2
Wall Shear Stress Values

Mode	Value	0 T	0.5 T	1 T	2 T	4 T	6 T
non-Newtonian	Maximum	0.432	0.432	0.464	0.588	0.876	1.338
	Minimum	0.198	0.1984	0.198	0.127	-0.950	-1.63
	fluctuation range	0.117	0.117	0.133	0.230	0.913	1.485
Newtonian	Maximum	0.345	0.343	0.378	0.506	0.829	1.477
	Minimum	0.148	0.149	0.149	0.083	-0.974	-1.614
	Fluctuation range	0.098	0.098	0.115	0.211	0.901	1.546

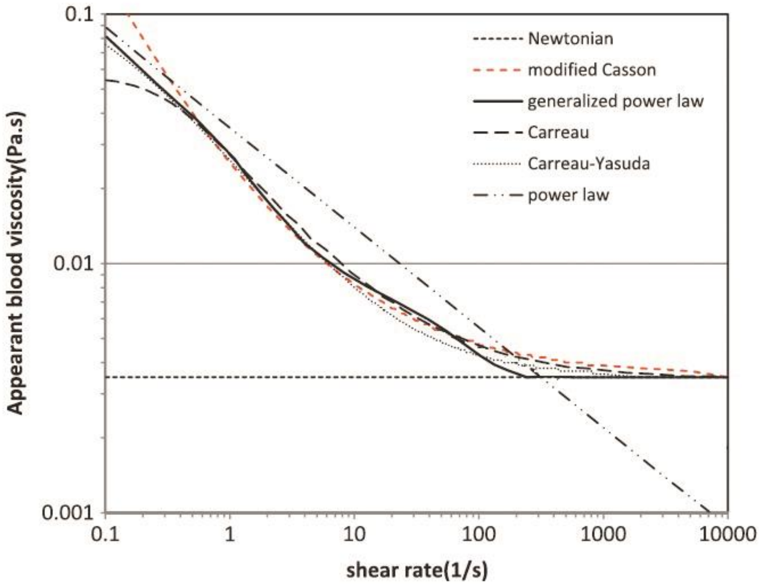

Fig. 11. Viscosity-shear rate diagram [15, 36]

5. Conclusion

Considering the importance and necessity of using simulations and engineering studies in the field of biomechanics as well as the effect of the magnetic field on the blood flow, many investigations have been done in this regard in recent decades [41-51]. Until the last decade, no significant progress was made in treatment by applying magnetic fields to target a specific site of the body through magnetic carrier particles. One of the most important reasons for the lack of rapid development of this method was to optimize the external magnetic field, to find the best properties for the Ferrofluid injected into the blood and to concentrate the required magnetic field in a small area of the body [52-54]. In this present work, a magnet with different field intensity was used to study the effect of the use of external magnetic field on the blood flow and the governing equations on the fluid and magnet were solved using the COMSOL Finite Element Software. The results of simulations were presented in two Newtonian and non-Newtonian fluid modes and hemodynamic parameters of blood pressure and WSS were investigated. From the results it was observed that there is no justification for the magnetic field of 6 T and more research is needed in this regard.

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