Final Report of Research Project

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Project title:

Modeling of Biofluid flow and Heat transfer through a coronary artery with a non-uniform Magnetic field

Submitted by:

Sorowar Hossain
Research Fellow
Department of Applied Mathematics
University of Dhaka

Supervised by:

Dr. S. M. Chapal Hossain
Assistant Professor
Department of Applied Mathematics
University of Dhaka

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Director
Bose Centre for Advanced
Study & Re...
Dhaka Univ...

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Abstract

Coronary artery disease (CAD) is one of the top causes of death around the world. Bypass graft surgery, which helps to get the blood flowing again, is a good way to treat this disease. This study examined and constructed a 3D computer model of a stenosed coronary artery bypass graft, including an investigation of the blood flow dynamics and hemodynamic parameters, to approach a realistic condition, inlet pulse, and non-Newtonian behavior. In the current simulations, the host artery is occluded with varying degrees of severity, such as 0%, 30%, 70% area stenosis. In addition, a modified multiphase flow model is implemented to gain a better understanding of bio-nanofluid (blood including Fe_3O_4 nanoparticles) flow regimes through a stenosed artery. Finally, A non-uniform magnetic field will use to eliminate the vorticity in a plaque region while also targeting nanoparticles in a specific region.

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1 Introduction

The blood rheology in the coronary artery is referred to as biofluid flow or hemodynamic flow in a coronary artery. It is important to know the blood flow dynamics in human body. Specially, to know the flow phenomenon of coronary artery. For this context, Mehmood et al. [1] studied of an unsteady 2D and axisymmetric flow of blood in an infected coronary arterial segment with adaptable walls. Yilmaz et al. [2] modeled pulsatile non-Newtonian blood circulation in coronary artery using a classical Eulerian multiphase method. Also, Ha and Lie [3] examines the impacts of pulsatile swirling inlet flows in a stenosed artery. When coronary arteries are narrowed, blood flows through them in a pulsing way could cause a change from a laminar flow to a turbulent flow [4]. Mahalingam et al. [4] showed that when there is fifty percent stenosis, turbulent flow starts. A similar study was done by Carvalho et al. [5]. In the other study, Misra et al. [6] used the Herschel-Bulkley fluid model to analyze the non-Newtonian characteristics of blood flow by putting a catheter into a stenotic artery. Also, to characterize the non-Newtonian blood flow, Guerciotti et al. [7] used unsteady computational modeling based on the finite element method and the Carreau-Yasuda model. Many researchers found that the non-Newtonian blood flow method best describes stenotic artery blood rheology[7, 8].

In modern world most of the people died all over the world due to the different types of heart disease[9]. The coronary arterial diseases such as coronary stenosis is one of them. Every year, 17.8 million people died globally and 610000 people died in united states for this disease [10]. Atherosclerosis is typically the cause of coronary artery stenosis [11]. It arises because of many hemodynamic parameters such as wall shear stress (WSS), oscillatory shear index (OSI), and excessive coronary artery curvatures, deposition of low - density lipoprotein, lipid, and other materials on the artery wall over time [12-14]. Many researchers estimated the hemodynamic parameters, curvatures of the artery, and bifurcation angle and showed the effect of these parameters on arterial diseases [4, 6, 15-21].

Nowadays, the impact of magnetic field on biofluid flow is a very interesting topic for researchers. Because magnetic field being used to treat many diseases such as cancer, stroke, arterial occlusion, atherosclerosis etc, also used for separation of cell, drug delivery for fixed destination [22, 23]. For this purpose, many researchers studied the effect of the external magnetic field on the biofluid flow and heat transfer through a stenosed artery [24-42]. Teimouri et al. [42] showed that the effect of a magnetic field on blood flow goes up when the percentage of blocked arteries goes down. To examine the dynamic characteristics of heat transfer in blood flow through a bifurcated artery in stenotic circumstances Jamali and Ismail [27] demonstrated that the presence of stenosis with a little variation in stenosis form and Reynolds number had significant effects on velocity, temperature profile, and reverse flow recirculation, as seen by negative flow close the artery wall. Using Casson fluid model and finite difference method(FDM) Sankar et al. [40] discovered that flow speed and flow rate decline as the Hartmann number increases, while the opposite is true for WSS and longitudinal resistance of the flow, also WSS and longitudinal resistance increase as the depth of stenosis increases. In another study, for Magneto Hydrodynamic flow (MHD) Majee and Shit [36] demonstrated that increasing the magnetic field magnitude up to 8 T did not cause harm to the artery wall, but the study is important for monitoring temperature increase during hyperthermic medication. In a separate investigation, Jamalabadi et al. [31] demonstrated that wall temperature regulation can maintain the optimal blood temperature range and that blockages of more than sixty percent result in a severe pressure reduction. Also demonstrated that when the Hartmann number increases, the blood vessel velocity decreases [31].

Magnetic drug delivery/targeting is so important for the treatment of arterial diseases. It also reduces the side effects of drug in the human body. Numerous researches were done by the Researcher's on this topic [28, 32, 43-47]. Teimouri et al. [42] showed that magnetic number, Mn=2.5 can be used as a secure field for Magnetic drug targeting in cases like stenotic artery. According to Das et al. [47] the blood temperature is reduced due to the dispersion of hybrid nanoparticles in contrast to copper blood or pure blood. This phenomenon occurred because the nanoparticles have a high thermal conductivity which removes the temperature quickly [28]. For a deeper understanding of bio-nanofluid fluid properties in an actual bifurcated artery, Hossain et al. [43] used a simplified multiphase flow model. They demonstrated that the pressure and vorticity profiles steadily grew and decreased with the presence of a magnetic field, and they determined the maximum pressure for a high current of I = 300 Ampere [43]. For magneto-nanoparticles (MNPs) delivery in stenosis, Varmazyar et al. [44] demonstrated that stenosis close to a magnetic field causes MNPs to become stuck nearest to the plaque, resulting in a decrease in the amount of drug available near tumorous tissue and a corresponding decrease in drug absorption. Before attempting to transport a medicine into a tissue, the presence of plaques should be noted [44]. The magnetic field enhanced density in the target location [46]. Bose and M. Banerjee [46] reveal that there exists an optimal regime of operating settings that maximizes the deposition of drug-carrying magnetic nanoparticles in a targeted region on the partially occluded artery wall which provided guidelines for the in vitro research of Magnetic drug delivery in stenotic blood arteries. The researchers came to the conclusion that a non-uniform magnetic field can change the secondary fluid flow in a big way which means this technique can be used to improve targeted drug delivery [33].

When serious coronary artery disease develops, a surgical technique called coronary artery bypass grafting (CABG) is performed to build new blood channels across stenotic coronary arteries to restore blood flow to the myocardium [48, 49]. CABG remains one of the most common surgical procedures performed around the world, despite the availability of other treatment options [50]. But it has many challenges which was described by many researchers [51-60]. Even though bypasses have a high success rate, restenosis can cause them to fail [52]. For this purpose, Guerciotti et al. [52] examined fluid dynamics in patient-specific geometries to investigate a potential correlation between coronary stenosis severity and graft failure risk. They demonstrated that low levels of coronary occlusion result in more perturbed fluid dynamics within the graft, leading to hemodynamic circumstances that may increase the risk of graft failure [52]. By analyzing the flow characteristics of bypass grafts, Ko et al. [60] give useful information for targeting the precise mechanisms responsible for restenosis in the host artery following bypass graft placement. In addition, Cacho et al. [58] devised a computational method for the modeling of traditional CABG operation and examined the impact of two significant clinical parameters on the artery and graft responses, namely incision length and insertion angle for a particular graft diameter. The most common reason for CABG to fail is stenosis of the distal anastomosis due to thrombosis and intimal hyperplasia [55]. Therefore, Ghista and Kabinejadian [55] examine the biomechanical aspects of improving the patency of coronary bypass grafts. Restenosis is also dependent on the angle of anastomosis. In this regard, Koksungnoen et al. [51] shown that a 45-degree anastomosis is optimal for treating coronary heart disease.

2 Methodology

2.1 Physical model and boundary condition

Our model is modified from Ko et al. [60]. The diameter of the host and bypass artery is $2mm$. The anastomotic angle was taken, $\theta = 45^{\circ}$. For our first assumption, we consider blood flow as laminar and blood is incompressible with density $\rho = 1000\ kg/m^3$ and dynamic viscosity, $\mu = 0.0035$

$Pa.s$ was used for a Newtonian blood flow. For non-Newtonian blood flow, we used Carreau-Yasuda model which taken form Kashyap et al. [20]. For pulsatile case, we used pulse inlet velocity, $v(t) = \frac{Q(t)}{A}$ where, A is inlet section area. Which is taken from the study of Impiombato et al. [59]. And, for discrete phase model, particle (Fe_3O_4 nanoparticles) properties was taken from Hossain et al.[43] and from Sundar et al. [61]. where the density of particle was $5200\ kg/m^3$. And volume concentration 0.2 was taken.

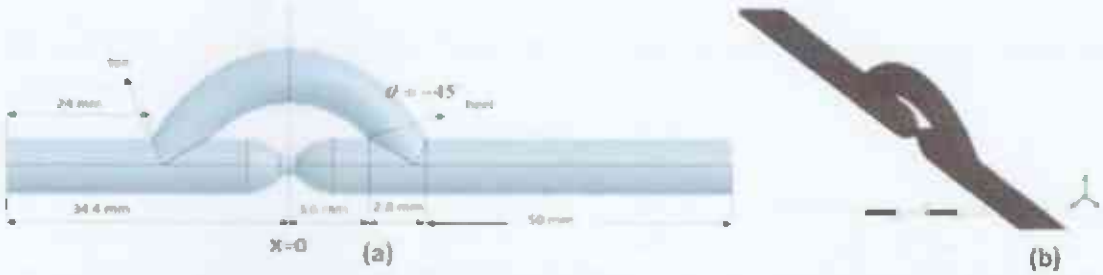


Fig.1 (a) Physical model (b) Isometric view of the computational mesh for the model

2.2 Governing equations

The Naviar-Stokes and the continuity equations represents the blood flow in a coronary artery with a bypass graft were modeled in a 3D form and written as follows:

Continuity equation:

$$\frac{\partial \rho}{\partial t} + \vec{\nabla} \cdot (\rho \vec{V}) = 0$$

Momentum equation:

$$\frac{\partial(\rho u)}{\partial t} + \vec{\nabla} \cdot (\rho u \vec{V}) = -\frac{\partial P}{\partial x} + \frac{\partial \sigma_x}{\partial x} + \frac{\partial \tau_{xy}}{\partial y} + \frac{\partial \tau_{xz}}{\partial z} + \rho f_x$$

$$\frac{\partial(\rho v)}{\partial t} + \vec{\nabla} \cdot (\rho v \vec{V}) = -\frac{\partial P}{\partial y} + \frac{\partial \sigma_y}{\partial y} + \frac{\partial \tau_{yx}}{\partial x} + \frac{\partial \tau_{yz}}{\partial z} + \rho f_y$$

$$\frac{\partial(\rho w)}{\partial t} + \vec{\nabla} \cdot (\rho w \vec{V}) = -\frac{\partial P}{\partial z} + \frac{\partial \sigma_z}{\partial z} + \frac{\partial \tau_{zx}}{\partial x} + \frac{\partial \tau_{zy}}{\partial y} + \rho f_z$$

where, $\vec{\nabla} = \frac{\partial}{\partial x}\hat{i} + \frac{\partial}{\partial y}\hat{j} + \frac{\partial}{\partial z}\hat{k}$

Where ρ is blood density, u, v, w is blood velocity in an x, y, z direction, $\vec{V}$ is the velocity vector, t is time, p is pressure, τ is shear stress, f is body force, σ is normal stress.

Shear stress for the Newtonian fluid is, $\tau_{ij} = -\mu\left[\frac{\partial u_i}{\partial u_j} + \frac{\partial u_j}{\partial u_i}\right]$, μ is dynamic viscosity.

For the non-Newtonian blood flow, the viscosity of fluid is shear dependent and described as:

$$\eta\left(\dot{\gamma}\right)=\eta_{\infty}+(\eta_{0}-\eta_{\infty})\left[1+\lambda\left(\dot{\gamma}\right)^{a}\right]^{\left(\frac{n-1}{a}\right)}$$ (Carreau-Yasuda model taken form Kashyap et al. [20])

Where, η_0 is the zero shear viscosity, η_∞ is the infinite shear viscosity, λ is a characteristic relaxation time, n is a flow index, $\dot{\gamma}$ is shear rate, α is Yasuda exponent. Flow rate, for pulsatile case is: $Q(t)=Q_0+\sum_{i=1}^{5}Q_i\cos(k\omega t-\varphi_i)$, where $\omega=\frac{2\pi}{T}$, T is cardiac cycle period $T=1.68s$, $Q_0=65.07ml/\min$ is average inlet flow rate, Q_i amplitude and φ_i is phase angle. These data taken from Impiombato et al. [59].

2.3 Numerical Procedure

The Navier-Stokes equation and associated boundary conditions were numerically simulated using the finite volume approach in this study. To describe the blood flow phenomenon within the coronary artery bypass graft, all computational methods were implemented utilizing Ansys fluent. The algorithm of the ANSYS ICEM CFD was used to generate the mesh and fig 1 (b) shows the structured grid used in this computation with 257161 nodes and 1274174 elements, where 70% stenosed artery was considered. The relative residual was set at 1×10^{-3} to assure the accuracy of the solutions.

2.4 Model validation

In this study, the simulation results were checked against the numerical results from Impiombato et al. [59]. In this validation case, the bypassed artery that was 70% blocked was chosen. It was assumed that blood had the properties of a Newtonian fluid with a density of 1000 kg/m^3 and a viscosity of 0.0035 $Pa.s$. Fig. 2 shows the time-dependent inlet flow rate $Q(t)$ in each case compared to the other. It found the time-dependent inlet flow rate from Impiombato et al. [59] and the solution given was close.

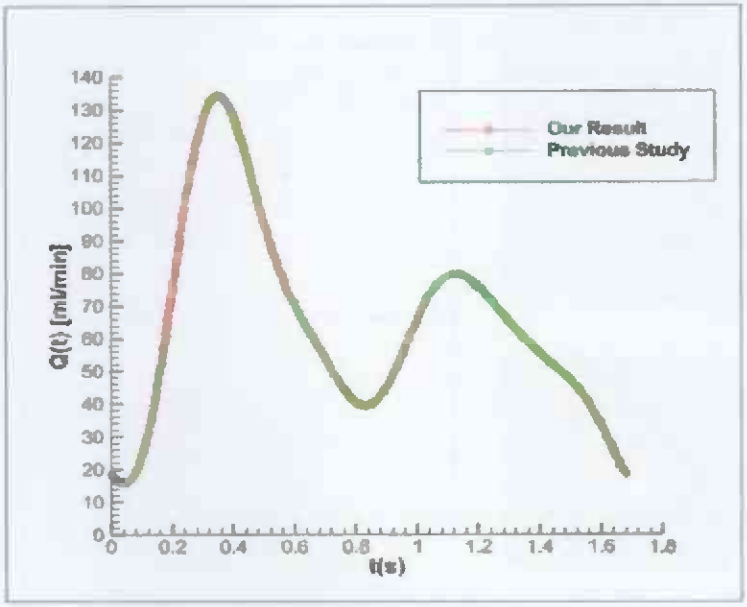
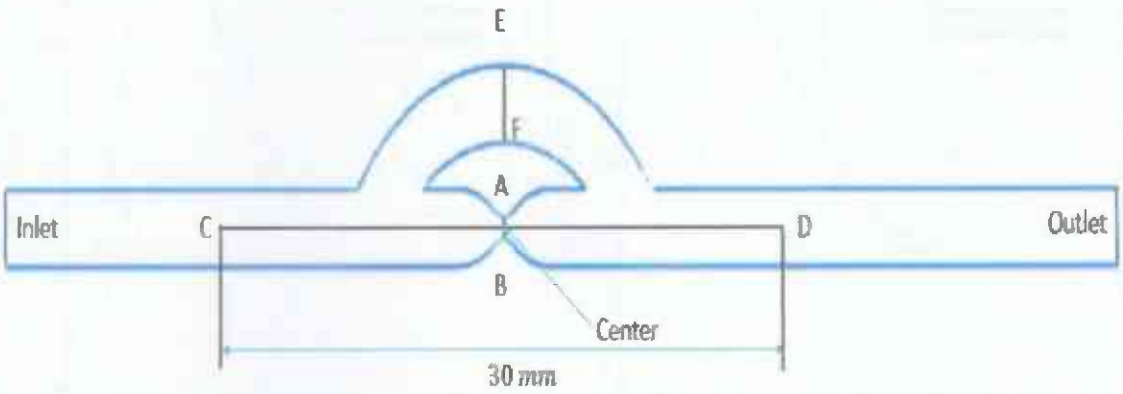


Fig 2. The comparison of the time-dependent inlet flow rate from Impiombato et al. [59] and our simulation

3 Results and discussion

In this stenosed CABG model, the effects of inlet flow (pulse and no pulse) on blood flow and wall shear stress were studied. The coronary artery blood flow phenomenon was examined. All scenarios used Navier-Stokes and continuity.



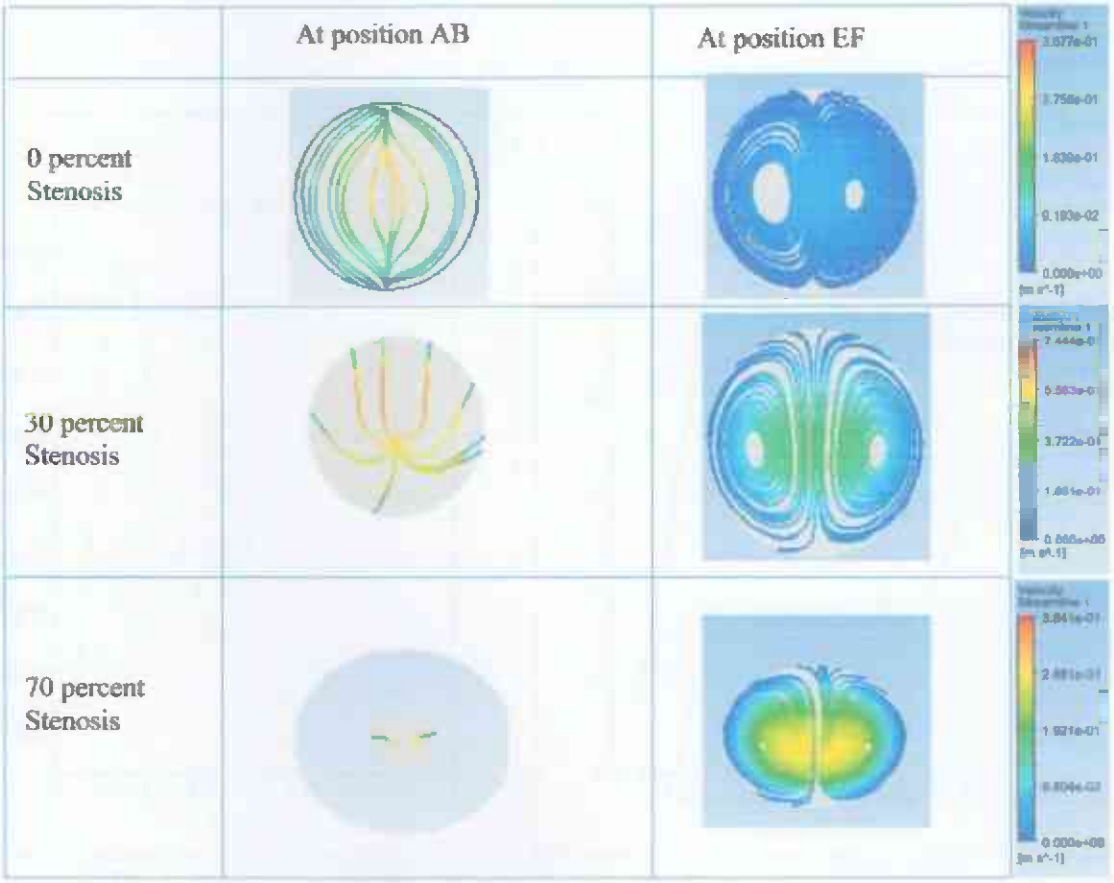


Fig 3. Velocity contour for different percentage of stenosis

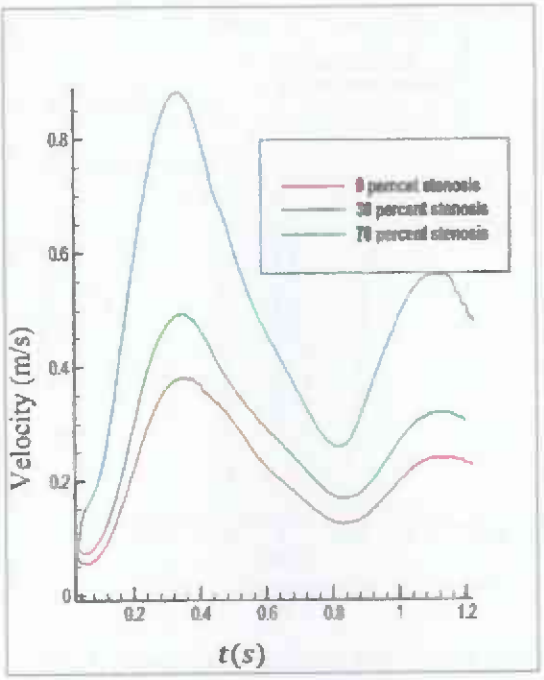


Fig 4 (a): velocity at region AB

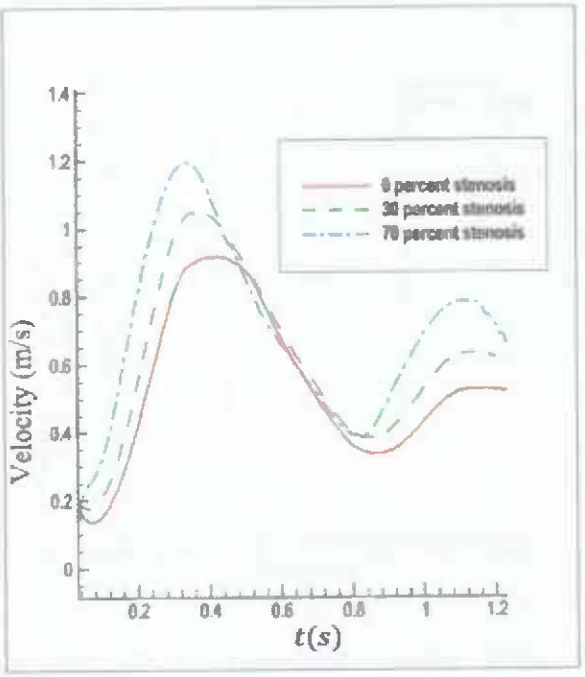


Fig 4(b): velocity at Center Point

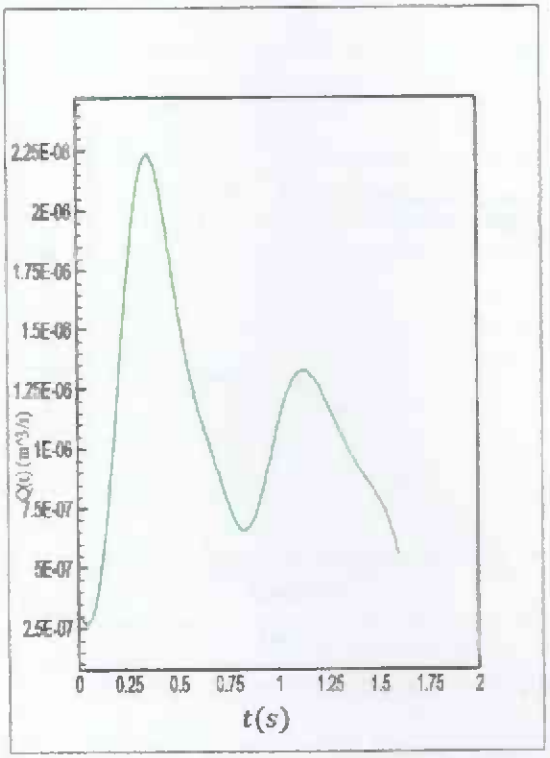


Fig 4(c): Flow rate at inlet

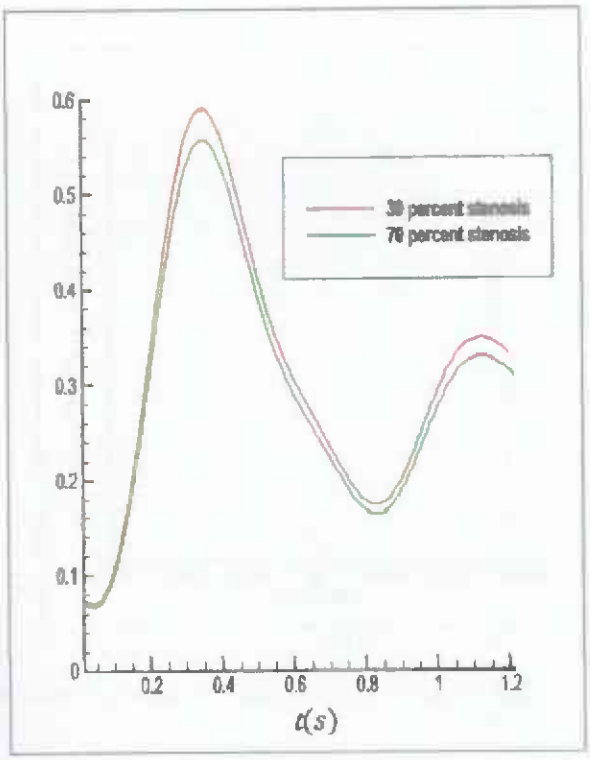


Fig 4 (d): velocity at inlet

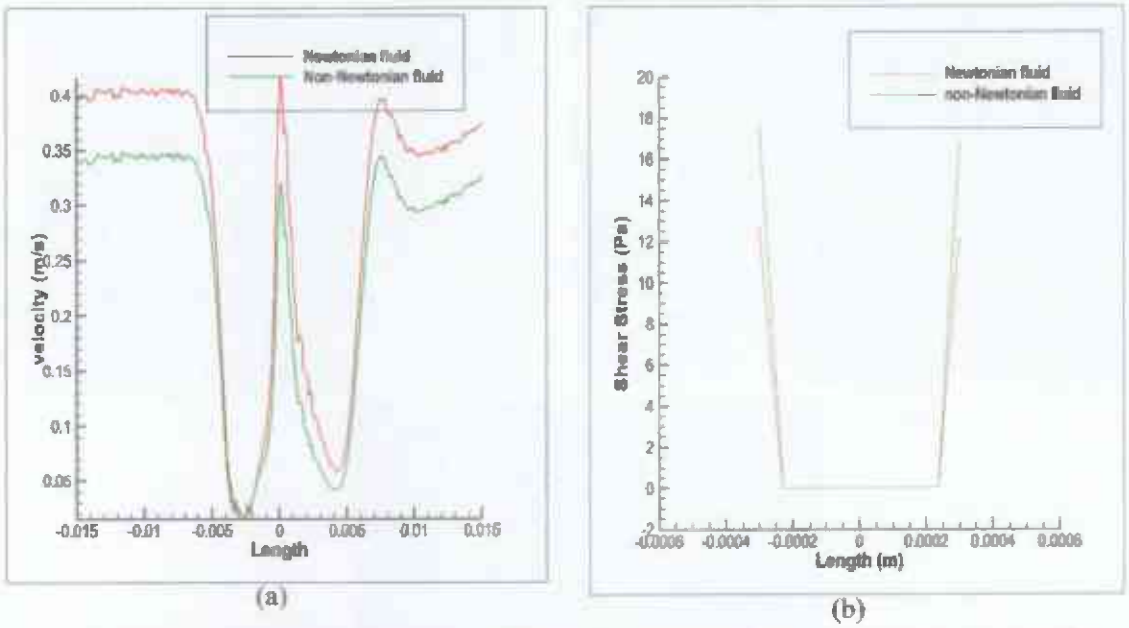


Fig 5: comparison between Newtonian and non-Newtonian fluid (a) velocity at region CD (b) shear stress at region AB

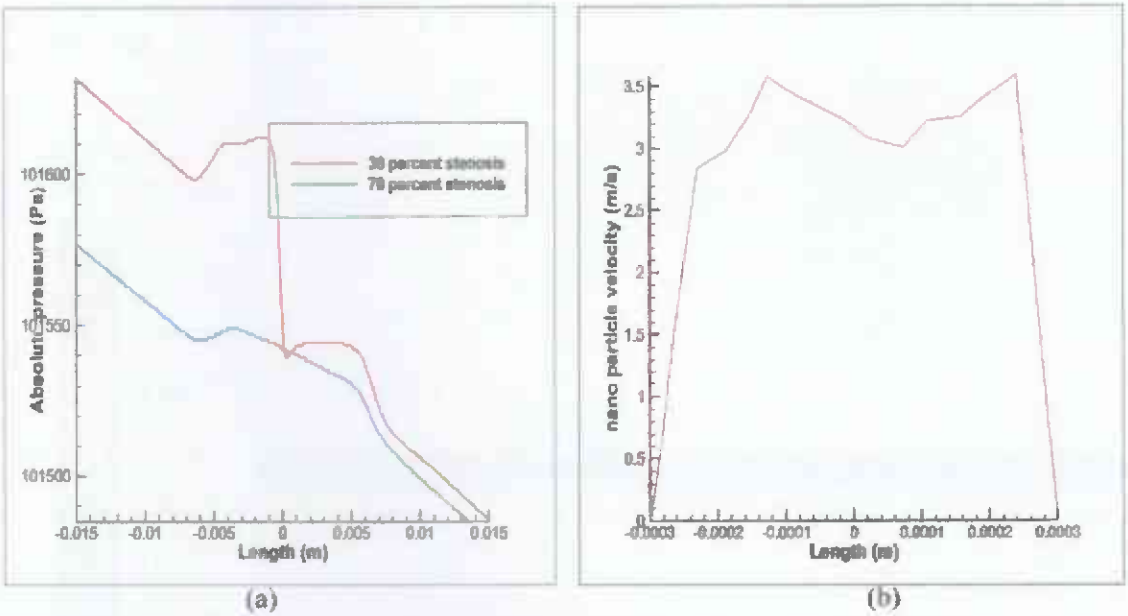


Fig 6: (a) comparison between Absolute pressure (b) nano-particle velocity at region AB

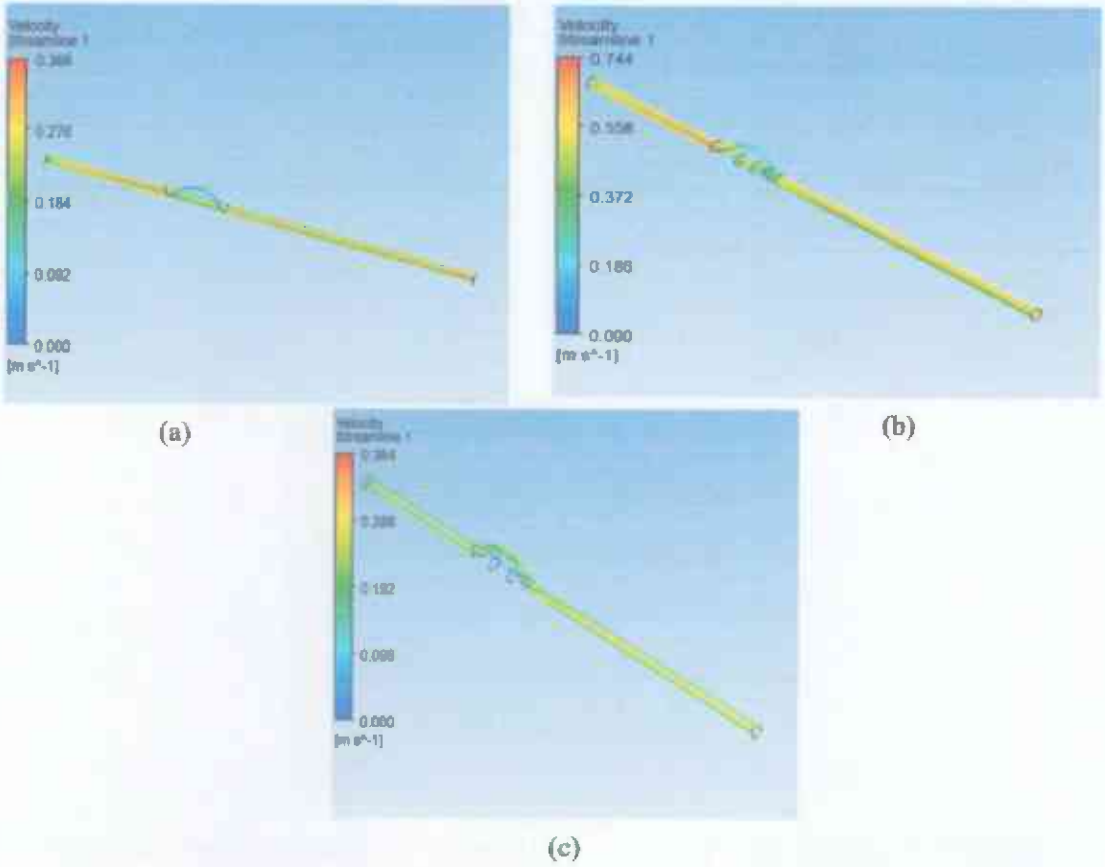


Fig 7: velocity streamline for (a) 0 percent stenosis (b) 30 percent stenosis (c) 70 percent stenosis

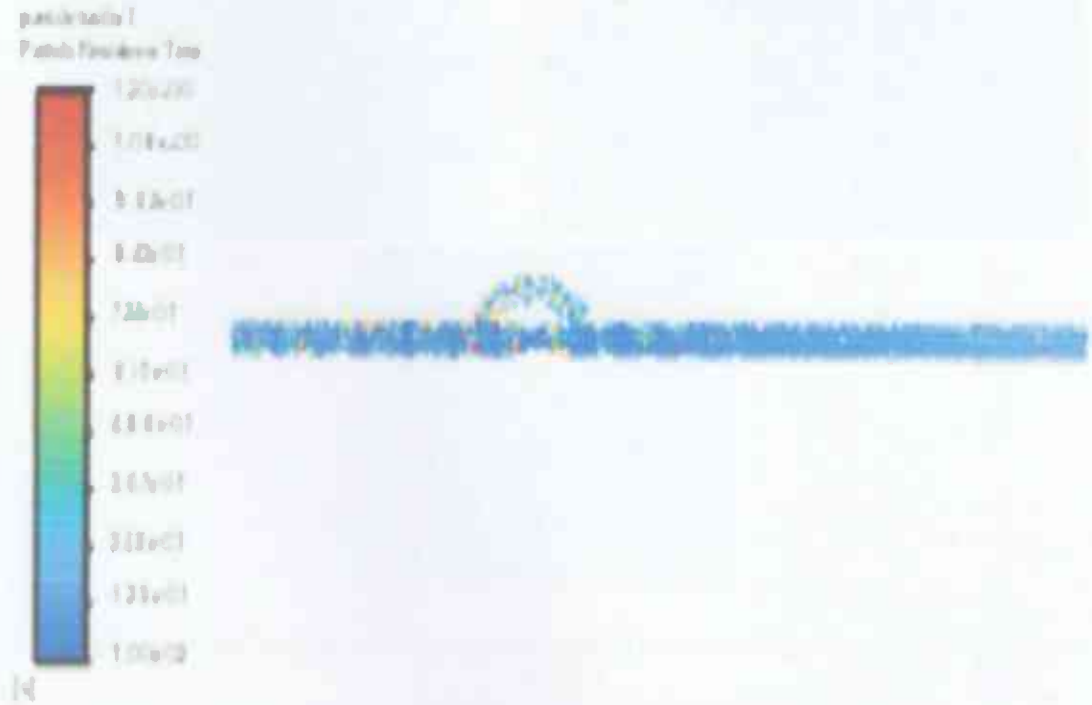


Fig 8: Particle deposition in our geometry

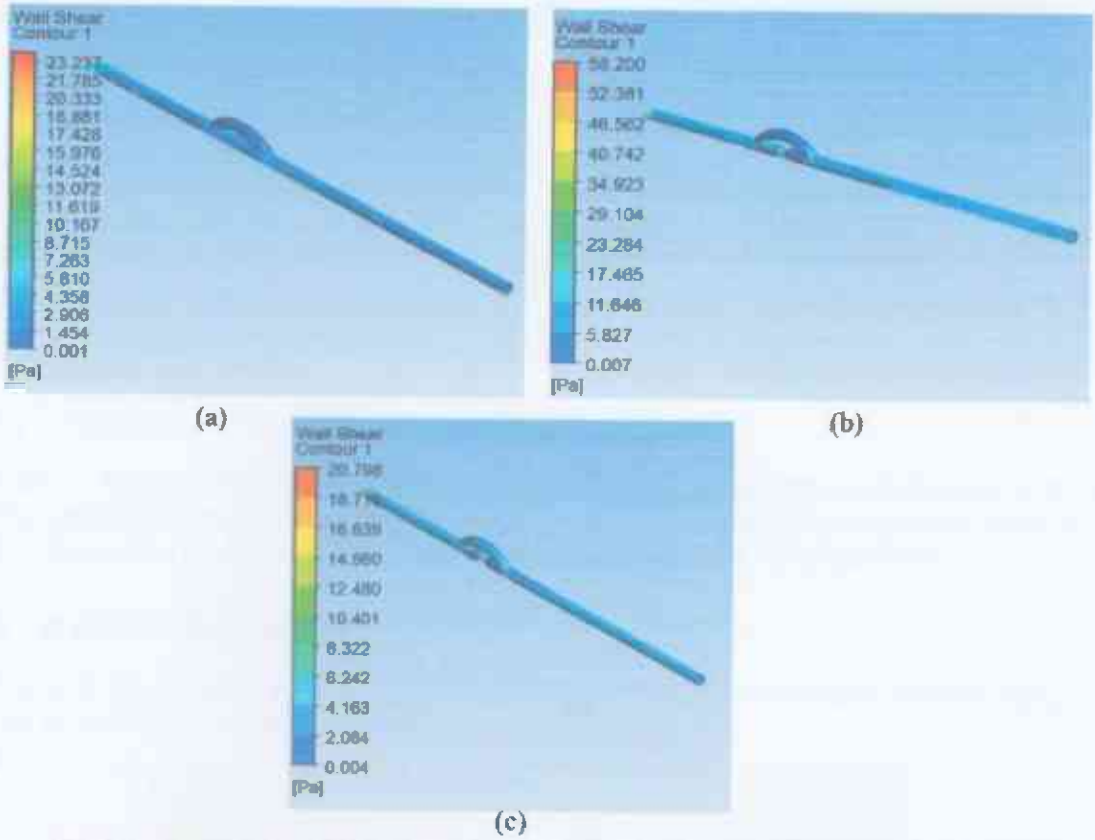


Fig 9: wall shear contour for (a) 0 percent stenosis (b) 30 percent stenosis (c) 70 percent stenosis

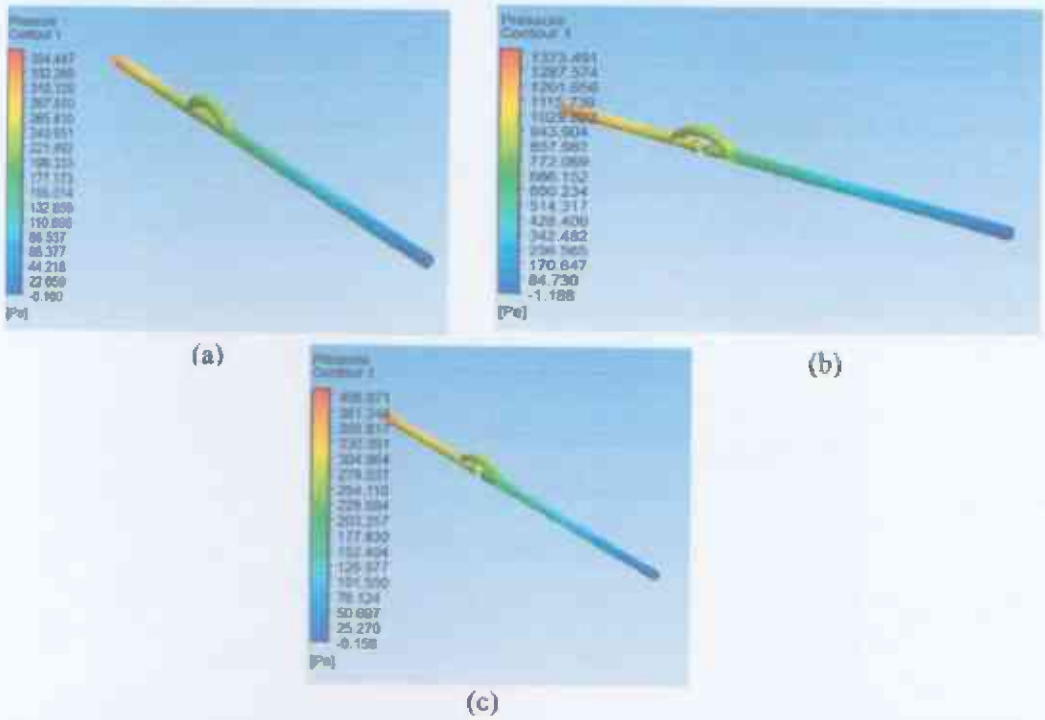


Fig 10: Pressure contour for (a) 0 percent stenosis (b) 30 percent stenosis (c) 70 percent stenosis

4 Conclusion

This research studied the flow behaviors in a 3D computer model of a stenosed artery bypass graft with varying percentage of stenosis, inlet flow situations with pulse situation. Fluid properties in the case of Newtonian and non-Newtonian blood. Also, we use discrete phase model to simulate the deposition of nano-fluid particles. The effects of these different parameters on the velocity, wall shear stress and pressure were systematically investigated. The fluid properties, Newtonian and non-Newtonian fluid gives almost same result. The inlet pulse flow situations showed an appropriate result more than others because of the similarity to a realistic systolic. The results presented here can be used as guidance for medical treatment and help to predict the outcome of the treatment.

4.1 Future research works

In future, A non-uniform magnetic field will use to eliminate the vorticity in a plaque region while also targeting nanoparticles in a specific region. And We will calculate hemodynamic factors and flow phenomena for different percentages of stenosis, such as 85%, 95%, and 100%.

5 Acknowledgment

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Fig 11: Participation as a Presenter at the A F Mujibur Rahman Bangladesh Mathematical Society National Mathematics Conference-2022

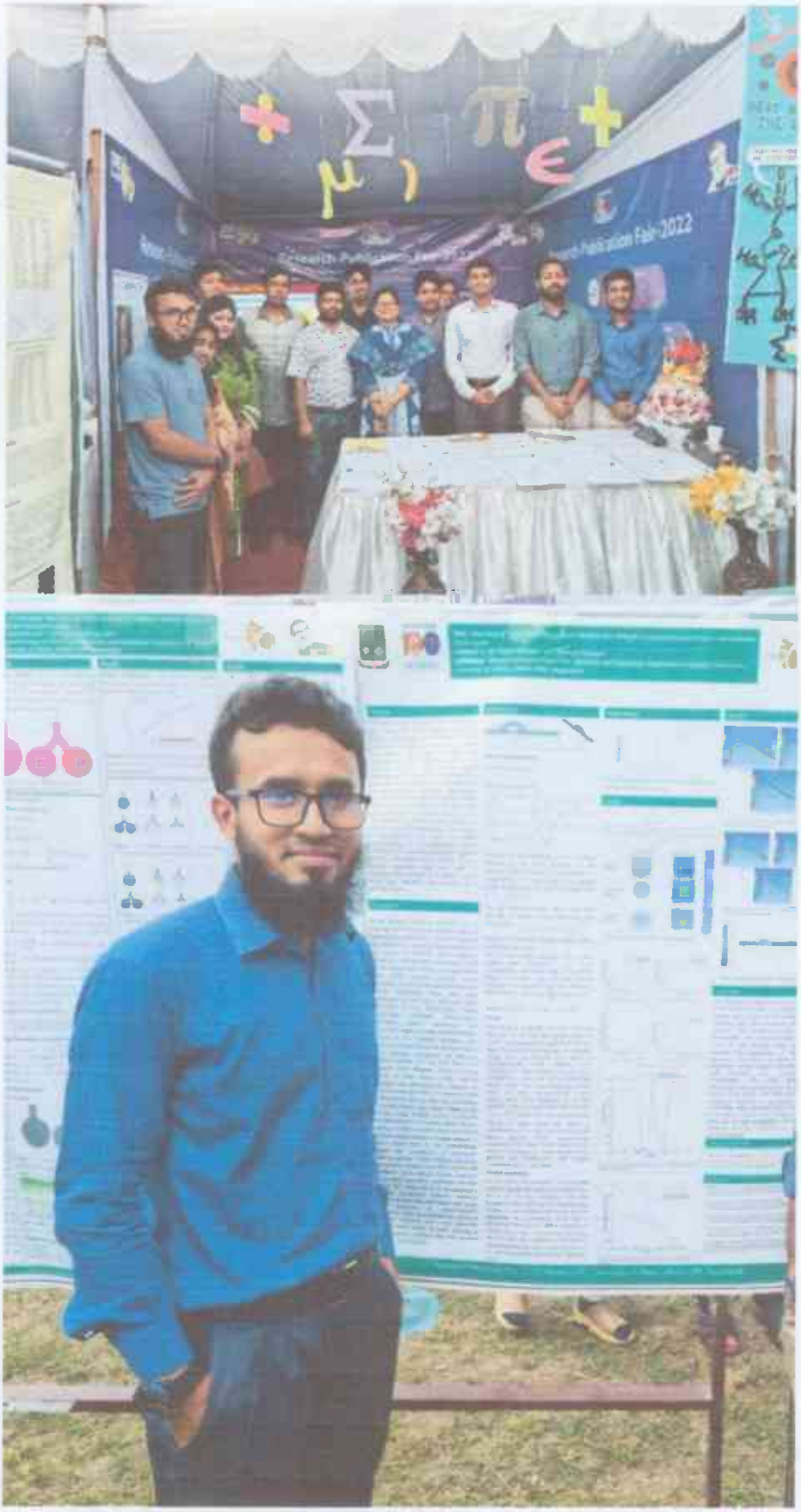


Fig 12: Participation as a Presenter at the Research Publication Fair, October 22--23, 2022, University of Dhaka



Fig 13: working at Research Lab

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