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Hemodynamic Characteristics in A Straight and Wavy Artery: A Numerical Study

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ABSTRACT

The aim of the present study is to investigate the hemodynamic characteristics through plain and wavy arteries with the help of numerical analysis using ANSYS FLUENT. The study is divided into two parts - steady flow through plain and wavy arteries and unsteady flow through plain and wavy arteries. Velocity profile for flow through a plain tube under steady-state shows no change in the pressure along with constant wall shear stress. In a wavy tube at a distance of 1.5d and 2.5d, there is the formation of small vortices. The pressure along the length of the artery keeps decreasing as the flow develops in accordance with Bernoulli's principle. For unsteady flow in a plain artery, the velocity decreases with time but there isn't much change. The wall shear stress becomes zero at the points where the flow separates because there is no momentum in the fluid. From the study, it was concluded that in a straight artery the streamline velocity remains constant through the length of the tube whereas in the wavy artery, streamline velocity shifts from the centerline.

Keywords: Hemodynamics, Pulsatile flow, Arteries, Numerical simulation, Wavy arteries.

1. INTRODUCTION

Flow-through arteries has been a topic of investigation for decades. In the field of biomedical science, it is essential to know how blood flows through the arteries as one can gauge potential diseases early on. Experimentally it is very difficult to analyze blood flow through the arteries, so developing a computational model to visualize blood flow is a need. Present study focuses on developing a computational model of blood flow through different geometries of the artery- straight and sinusoidal. Many researchers have investigated the hemodynamics under different circumstances and the following has been presented in this section.

Johnston et al. [1] investigated the shear stress pattern during specific points of a cardiac cycle. For the study, four separate right coronary arteries were simulated, and each one was analyzed using five non-Newtonian models and a Newtonian model. There is no substantial difference between the models at low velocity values. Regardless of the model utilized, similar patterns of wall shear stress were seen in each of the four arteries. It was concluded that the Newtonian model can be approximated as a good model for the computational analysis of the heart. Soulis et al. [2] investigated the distribution of wall shear stress along the left coronary artery tree using computational models. It was discovered that low levels of wall shear stress were detected along the arteries where they were branching. At flow dividers, high values of wall shear stress were found. The distribution of wall shear stress was discovered to be crucial in determining atherosclerosis. He and Ku [3] investigated the effect of pulsatile motion in the left coronary artery with bifurcations. Reverse velocities were present during early systole due to the accelerating nature of the waveform. It was also seen that the bifurcations present in the artery causes more skewing than the myocardial surface curvature. Nevertheless, curvature of the arteries created strong waves of secondary flows which was highlighted by the circumferential velocity plots. Weydahl and Moore [4] studied how blood flow patterns can help in determining the atherosclerosis formation. This could be easily determined in an artery bifurcation model. The inner walls of the artery showed low values of shear rate. The opposite effect was seen on the outer walls of the artery. It was concluded that the geometrical parameters can be essential in determining atherogenesis as these are formed in regions of low mean and oscillating wall shear. Santamarina et al. [5] simulated an artery as a tube with a varying radius of curvature. The velocity of blood flow at the entrance of the artery was set to Reynolds number of 300. In the first readings, the curvature of the arteries was held constant and in the second readings, the curvature varied as a sinusoidal wave. The wall shear rates fluctuated as much as 52 percent of the static mean wall shear rate when the artery curvature was considered to be constant. Schilt et al. [6] analyzed the effects of varying curvatures on the velocity profiles of blood flow in the arteries. To investigate these parameters an in vitro model was created to simulate the flow as close as possible to the human heart. A stepper motor and a carriage were used to change the artery's radius of curvature. The maximum skewing of the velocity was seen when the radius of curvature shifted from a lower to a higher value. When the carriage moved obliquely, the change in skewing was greater than when it went perpendicularly. Moore et al. [7] looked at how blood flow patterns can aid in the early detection of cardiovascular illnesses. Blood flow was studied throughout both systole and diastole phases of the cardiac cycle. The wall shear rate showed considerable fluctuations during these phases. Most changes in parameters were observed during the systolic phase when the pressure gradient is minimal but the movement is strongest. The liquid stream in the human left foremost plummeting coronary vein was analyzed by Ramaswamy et al. [8]. The investigation's objective was to observe the speed profiles and dissemination

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of divider shear pressure in the stenosis zone. The conduit divider movement was found to adjust the fleeting mean divider shear pressure and oscillatory shear file, recommending that observing these measurements could give a more practical chance to the atherosclerosis forecast. The flow patterns and spatial distribution of atherosclerotic lesions in human arteries were examined by Asakura and Karino [9]. Five separate transparent human coronary artery trees were recreated postmortem to assess fluid flow in the arteries. The outer walls of the daughter's vessels were discovered to have areas of atherosclerosis and wall thickenings. As a result, these locations became recirculation zones with low-velocity values corresponding to low shear stress. Chiastra et al. [10] investigated artery hemodynamics in the presence of stents. Recreated are two cases of diseased left anterior descending coronary arteries. The regions closest to the stents were shown to be more prone to restenosis, as evidenced by the values of local wall shear stress and relative residence time. Microstructures with helical recirculating helices can also be seen downstream of the stent. The present computational study is focused on the effect of the waviness of arteries on blood flow behavior. Wall shear stress and velocity profiles are compared for straight and wavy tubes.

2. GEOMETRY OF THE ARTERIES

In the present study, a comparison has been made between a straight artery and a sinusoidal artery. Geometric details for the plane and wavy artery model are shown in Fig. 1. The diameter of the tube is taken as 8 mm. The effect of the geometries of the two arteries on the hemodynamic parameters has been presented. In the first case, the artery is considered as a straight tube whereas in the second case artery is considered as a wavy tube with sinusoidal nature.

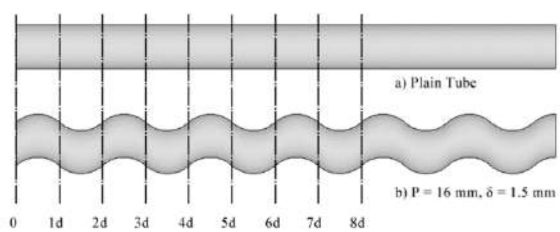

Figure 1: Geometric Details of the straight and wavy artery

3. METHODOLOGY

Mathematical Model

In the current investigation, the geometry of the model was displayed utilizing SOLIDWORKS, and meshing was finished utilizing ANSYS 2020 R1. Blood flow in the arteries has been considered as steady/unsteady, pulsatile, isothermal, incompressible, and Newtonian. Blood density and viscosity are estimated at 1050 kg/m3 and 0.0035 Pa-s, respectively. Flow velocity is considered as 0.149 m/s which gives the Reynolds number of the flow as 360.

Mesh Topology

To get an accurate solution it is essential to select a proper mesh size. This can be achieved by carrying out a grid independence test on the model. The ANSYS Meshing tool was being used to mesh the data. Simulations for different mesh sizes were run and it was found that the model became independent for a mesh with an element size of 0.55 mm and 0.33 million elements. For the simulations, linear element order was used. A tetrahedral element grid with five inflation layers at the walls was also used. Edge sizing is also chosen to refine the mesh at the boundaries, increasing the degrees of freedom in the areas of interest. The single tube had a count of around 0.25 million.

A pressure-based solver with coupled scheme along with a Least Square cell-based gradient was used for the simulations. SST k- turbulence model was used in the study, considering its stability and precision. Reynolds Averaged Navier Stokes equations (RANS) are used in the present study. The governing equations are as follows.

Continuity equation:

$$\frac{\partial u_i}{\partial x_i} = 0 \tag{1}$$

Momentum equation:

$$\frac{\partial u_i u_j}{\partial x_i} = -\frac{\partial \rho}{\rho \partial x_i} + \frac{\partial}{\partial x_j}\left((v + v_t)\left(\frac{\partial u_j}{\partial x_i} + \frac{\partial u_i}{\partial x_j}\right)\right) \tag{2}$$

Energy equation:

$$\frac{\partial u_i T}{\partial x_i} = \rho \frac{\partial}{\partial x_i}\left(\left(\frac{v}{Pr} + \frac{v_t}{Pr_t}\right)\frac{\partial T}{\partial x_i}\right) \tag{3}$$

Turbulent kinetic energy k equation:

$$\frac{\partial u_i k}{\partial x_i} = \frac{\partial}{\partial x_i}\left(\left(v + \frac{v_t}{\sigma_k}\right)\frac{\partial k}{\partial x_i}\right) + \Gamma - \varepsilon \tag{4}$$

Rate of energy dissipation ε equation:

$$\frac{\partial \rho u_j}{\partial x_j} = \frac{\partial}{\partial x_i}\left(\left(v + \frac{v_t}{\sigma_k}\right)\frac{\partial \varepsilon}{\partial x_i}\right) + C_1 \Gamma \varepsilon - C_2 \frac{\varepsilon^2}{k + \sqrt{v\varepsilon}} \tag{5}$$

Where Γ denotes the production rate of k and is given by:

$$\Gamma = -\underline{u_i u_j}\frac{\partial u_i}{\partial x_i} = v_t\left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i}\right)\frac{\partial u_i}{\partial x_i} \tag{6}$$

$$v_t = C_\mu \frac{k^2}{\varepsilon} \tag{7}$$

The coefficients in the k- ε turbulence model is given as follows:

$$C_1 = max\left[0.43\frac{\mu}{(\mu_t + 5)}\right], C_2 = 1.0, \sigma_k = 1.0, \sigma_\varepsilon = 1.2 \tag{8}$$

To achieve the realizability effect C_\mu is no longer constant, but a function of turbulence fields, mean strain, and rotation rates.

Boundary Conditions

Limit conditions were set like that of Gupta et al. [11]. The limit conditions were characterized for both consistent states just as pulsatile recreation. In both of these streams, blood was considered a Newtonian liquid. Velocity inlet was used at the