Blood Flow in Human Arterial System-A Review

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Abstract

The blood flow in human arterial system can be considered as a fluid dynamics problem. Simulation of blood flow in the arterial network system will provide a better understanding of the physiology of human body. Hence, hemodynamics play an important role in the development and progression of arterial stenosis, leading to the malfunctioning of cardiovascular system. Simulation studies of blood flow in the diseased condition can diagnose the health problem easily and also have many applications in the areas such as surgical planning and design of medical devices. This paper presents a review on the existing scenario of the simulation studies of blood flow, starting with a brief overview of the structure and function of arteries and veins followed by a discussion on pressure wave propagation, blood flow models and Fluid Structure Interaction in the arterial system.

1. Introduction

The main function of cardiovascular system is to transport nutrient and waste throughout the body. The majority of deaths reported in the developed countries result from cardiovascular diseases. Earlier, most of the cardiovascular disease affects the aged group, but, that situation is different now. There are several other risk factors for heart diseases like age, gender, use of tobacco, high blood pressure and cholesterol etc., causing the development of stenosis. Hemorheology is an area of science concerned with the blood flow and its interaction with the blood vessel through which the flow occurs. The human blood circulatory system provides essential substances such as nutrients and oxygen to the cells and transports metabolic waste products away from the same cells.
composed of blood cells suspended in blood plasma. Plasma, which constitutes 55% of blood fluid, is mostly water (92% by volume), and contains dissipated proteins, glucose, mineral ions, hormones and blood cells themselves. The blood cells are mainly red blood cells (also called RBCs or erythrocytes) and white blood cells, including leukocytes and platelets. The red blood cells are small semisolid particles, increase the viscosity of blood and will affect the behaviour of fluid. It has been pointed out that plasma behaves as a Newtonian fluid [1] whereas whole blood, shows non-Newtonian character [2].

Healthy blood vessels are complex in structure. There are three major types of blood vessels: the arteries, which carry the blood away from the heart at higher physiologic pressures, the capillaries, which enable the actual exchange of water and chemicals between the blood and the tissues, and the veins, which carry blood from the capillaries back toward the heart at lower physiologic pressures. Because of their different roles, their structures and wall constituents are also different. The wall of large blood vessels has a circumferentially layered structure. The most important layers are intima, media, and adventitia. The internal intima, composed of the endothelium cell. The media, which is a layered one, is responsible for most of the vessel mechanical properties. The outer layer is adventitia. Veins have a different layered configuration than arteries. They have a thinner wall, a less elastic media, and a thicker collagenous adventitia. The walls of the capillaries are extremely thin, constructed of single-layer, highly permeable endothelial cells [3].

The important feature of the arterial blood flow is its pulsatile nature. The left ventricle chamber of heart ejects the blood intermittently to the whole body system [4,5]. Normally arterial flow is considered as a laminar flow. But the development of stenosis, stiffening of arterial wall etc., will cause turbulence and reduce the required blood flow, leads to the malfunctioning of various organs. Hence detailed knowledge of blood flow is a fundamental key concept in the detection of arterial diseases [6,7]. The principal quantities which describe the blood flow are the flow velocity \( u \) and pressure \( p \) [8]. But, in the fluid-structure interaction problems, the displacement of the vessel wall due to the action of the flow field is also to be considered.

2. Simulation models

The wide span of topics such as blood pressure wave propagation, blood flow models, fluid structure interaction models etc. are covered in the literature review [10 - 40], will help the better understanding of blood flow models in humans. The related topics are discussed in the subsequent paragraphs.

2.1. Pressure wave propagation

Blood pressure is the pressure exerted by the circulating blood upon the walls of blood vessels of body. Several systems of body will help the regulation of blood pressure. In arteries, blood pressure is more, compared to the vein system of the human body. Sometimes due to some abnormalities, the blood pressure in the arteries may arise. Pressure waves are formed mainly due to the pumping action of heart. Simulation studies of pressure waves in the human system has very important role in diagnosing the diseased area of the human body. For the better understanding of the blood flow in arteries, pressure wave propagation through an initially stressed tube which carries a viscous, incompressible fluid has been considered [9]. In this model, fluid is assumed as Newtonian and the tube is taken as elastic and isotropic. Even though the arterial wall in thick wall concept gives more idealistic results, but in this study the wall is assumed as thin wall. But, this study excludes the non linear terms in the fluid flow. It is found that the longitudinal displacement to large pressure increments are very large compared with the observed longitudinal oscillation of arterial wall.

By assuming the fluid flow as laminar, the pressure wave propagation through a viscous fluid flowing in an elastic thick walled orthotropic tube is studied [10]. The arterial wall motion equation is modified for longitudinal displacement with longitudinal tethering of arterial wall. In this study the reflected wave concept is also considered. It is found that the blood flow rate is decreased as pulse propagates away from the heart. Cox [11] considered the propagation of harmonic pressure waves through a homogeneous Newtonian fluid flowing through a thick walled viscoelastic tube. The motion of the fluid is described by the linearized form of the Navier-Stokes equations. The
motion of the wall is explained by classical elastic theory. The solutions of the propagation of forced pressure waves are formed in terms of the finite series of harmonic terms. From the results it is found that the fluid impedance predicted by this model is smaller than the rigid tube model values.

All the above papers mainly used linearized Navier-Stoke equation for explaining blood flow in arteries. Because of the large dynamic storage effect of the arteries, the nonlinear convective acceleration terms of the Navier-Stokes equations have to be incorporated. Ling and Atabek [4] has been developed a theory for calculating the blood flow in large arteries by using the nonlinear terms of the Navier-Stokes equations as well as the nonlinear behaviour and large deformations of the arterial wall. In this study, axisymmetric flow is assumed. The theory discusses the velocity distribution and wall shear stress in terms of locally measured values of the pressure, pressure gradient and pressure-radius relation. The initial stress distribution in the arterial wall is recognized as non uniform and therefore constant initial stress assumption cannot be applied in such conditions [12]. An alternate approach is introduced by considering the nonlinear viscoelastic and orthotropic nature of arterial wall, blood is treated as Newtonian, viscous incompressible fluid [13]. In this study, a frequency equation with the help of equation of motion of the fluid and wall, together with the equation of continuity is derived and, computed the phase velocities with frequency corresponding to different transmural pressures. From the result, it is found that phase velocities increase with increase of transmural pressure.

A theoretical analysis is developed for studying the effect of initial stress in blood flow arteries using the wave propagation through an initially inflated cylindrical thick shell [14]. The assumptions taken for this study are variable initial stress, incompressible inviscid nature of fluid and the large static deformation of arterial wall. At this existing condition a small dynamic deformation is added. It is found that the speed of pressure waves increases with increasing inner pressure. By treating blood as a homogeneous, incompressible fluid, Wang and Tarbell [15] studied the non-linear behaviour of Newtonian fluid subjected to the oscillatory pressure gradient. It is also assumed that artery wall is isotropic, thin-walled, and elastic tube and its longitudinal wall motion is negligible. From this study, it is found that the nonlinear convective acceleration generate a finite mean pressure gradient when there is no mean flow in the arteries. The Method of characteristics is applied for calculating the pressure pulse propagation in large arteries [16]. The model is used to study the effects of reflected and absorbed waves. From the result, it is found that complex wave pattern is an important factor in the arterial hemodynamics.

Wave propagation through a fluid filled elastic tapered, thin walled long circular tube has been attempted by considering mean pressure as 100 mmHg [17]. The main assumption in this study is that the arteries are initially subjected to static deformation. But due to the blood flow through arteries, an additional dynamic deformation will occur. The governing equations of tube are taken as similar to the prestressed thin and tapered elastic tube. The equations are developed for static deformation and modified to incorporate the dynamic deformation concept. The fluid is treated as incompressible, non-viscous and one dimensional. The paper concluded that wave speed increases when the tube radius decreases. The wave speed in the human coronary artery is studied after analysing the same in aorta [18]. In this study, the coronary wave speed is derived from the single point method. The results showed that both coronary and aortic waves have good correlation.

Matthys et al., [19] developed an experimental setup to study about the pulse wave propagation in arterial system and is compared with the results of numerical analysis. From the result it is concluded that one dimensional models can also capture the wave propagation features. The paper is also mentioned about the importance of wall visco-elasticity. The wave intensity analysis is carried out for studying the waves in blood vessels [20] and applied in the coronary wave analysis [21].

2.2. Blood flow models

The unsteady entry blood flow in a 90° curved tube is numerically and experimentally investigated by comparing the Newtonian and non-Newtonian blood models [22]. For modelling purpose, non-Newtonian nature of blood flow is considered. Both numerical and experimental results are in good agreement. The steady blood flow through the four different right coronary arteries has been studied with the help of an angiograms data [23]. The wall shear stress distribution in the arteries is studied with the help of five different non Newtonian models and one Newtonian model. The general comparison of non-Newtonian models is as follows: In Power Law and Walburn–Schneck models, viscosity decreases due to increase in strain up to 226.5 s⁻¹. But, in Casson and Carreau model,
viscosity tends to a limiting value at high shear rate. In some special cases of Generalised Power Law models can incorporate Power Law model, Newtonian model and Casson model. In all the models, blood is assumed as an incompressible fluid with a density of 1050kg/m³, governed by Navier-Stoke equations and continuity equations. The artery is considered as a rigid tube. The equations are solved by using finite volume method as implemented in package CFD-ACE. The flow equations are solved with the help of boundary conditions. At the outlet, stress free conditions are applied and at boundary wall, no slip condition is assumed. The velocity profile is assumed to be paraboloidal shape and the centre line velocity values taken are 0.02, 0.05, 0.1, 0.2, 0.5 and 1 m/s. Wall shear stress distribution of the right coronary artery is plotted for various models for selected inlet velocities 0.02, 0.2, and 1m/s and also for Generalised power law model. The pattern of wall shear stress in a particular artery for a particular inlet velocity is same for all non Newtonian and Newtonian models. All the model results shows the same trend throughout. Based on the changes of inlet velocity, the magnitude of wall shear stress varies. The magnitude difference of wall shear stress is significant at low inlet velocity. The power law models over estimates wall shear stress at low inlet velocities and underestimates it at high inlet velocities. The Walburn-Schneck model under estimates wall shear stress at high inlet velocities and the Newtonian model under estimate wall shear stress at low inlet velocities. Due to the above reasons the power law and Walburn Schneck is not suitable for the blood viscosity modelling relative to other models. The Newtonian model is more suitable in mid range to high shear rate. The Generalised model (non-Newtonian model) gives better result in wall shear stress than Newtonian model and mainly important at low inlet velocities.

A mathematical model is developed for the non-Newtonian blood flow through a flexible tapered artery with stenosis [24]. The non-Newtonian blood characteristics is explained by generalised power law model. The blood vessel is modelled as thin elastic tube with circular cross section. The blood flow is assumed as two dimensional, unsteady, axisymmetric and fully developed flow. The continuity and momentum equations in cylindrical coordinate system are used. Finite central difference scheme is used to solve the unsteady non linear Navier-Stoke equation. This study also focused on the nonlinearity and wall distensibility concepts. The axial velocity and radial velocity profiles in the stenosis region are also plotted. It is found that, the velocity of the blood flow is affected by the presence of stenosis, tapering of arteries and due to the steepening of stenosis. The blood flow experienced high resistance in arterial construction in rigid wall concept.

Johnston et al., modelled the pulsatile blood flow through the four different right coronary arteries [25]. The data taken for this study is the same as given in [23]. A non-Newtonian blood model (the Generalised Power Law), as well as the usual Newtonian model of blood viscosity, is used to study the wall shear stress in each of these arteries over the entire cardiac cycle and the results are compared. The results mentioned are in terms of the wall shear stress distribution, non-Newtonian importance factor and particle path simulation. It is found that, the wall shear stress is low at the inlet part and increasing towards the outlet due to the tapering effect of arteries. Based on the wall shear stress distribution study it is found that Newtonian blood flow modelling is a good approximation.

The blood flow through the vertebral arteries in human system has been studied by selecting the geometry with one inlet and six outlet [26]. The flow parameter study mainly focused on the region where the vertebral arteries (2-6mm diameter) joined to form Basilar artery (3-8 mm diameter). Twenty five similar cases were selected and studied the simulation of blood flow. For the simulation purposes, the vessel wall is considered as rigid and assumed a constant blood density of 1055kg/m³. The Power law non-Newtonian numerical model is applied in this study. At the outlet and inlet, the velocity variation and pressure variation versus time is explained based on the Fourier series. By considering the blood flow as turbulent, shear stress transport model is used for the analysis. For this purpose, the model run in steady conditions and parameters for turbulent modelling are taken from steady conditions. The ANSYS CFXv.10.0 solver is used and the velocity distribution along the diameter of the left (2-6mm diameter) and the right (2-6mm diameter) vertebral arteries for one geometrical configuration at different instants of the heart operation cycle (t = 0.15s, 0.2s, 0.25s, 0.30s) is plotted. It is found that, at lower diameter, the flow profile is similar to laminar flow but in large diameter, the profile shows some deformation. The diameter combinations of left and right arteries showed various velocity profile at three time instants (t = 0.15s, 0.20s, 0.25s ) of heart operation to get a clear idea for velocity profile in both right and left arteries. From the result, it is found that individual configuration and diameter of arteries have an influence on the flow.

The unsteady blood flow in large arteries is numerically simulated using three dimensional and one dimensional models [27]. The artery network is considered as a rigid network system. The one dimensional flow
equation is solved by Galerkin method. The blood is assumed as incompressible Newtonian fluid with constant density and viscosity. The original MRI images of two arterial network (one incomplete and one complete) are used for modelling. The mesh is created using commercial software Gridgen. The simulation is carried out in second order semi implicit time splitting scheme. At the inlet, the flow rate condition is applied. At all the outlets, constant pressure condition are applied. Pressure drop and flow rates in the terminal branches and inlet are compared in one dimensional and three dimensional rigid modelling. In one dimensional model, the pipe diameter is constant everywhere and also the angle of bifurcation is not included in the model. Due to this, in flow rate comparison results, some differences are observed between the two models. The rigid model concept in one dimensional is replaced by flexible wall model and compared this result with rigid wall three dimensional model. In this model, at the outlet, time dependent pressure is applied. The fluctuation of pressure and flowrate in three dimensional rigid wall is more compared to one dimensional flexible wall. Hence, the elasticity properties of artery wall are also to be considered in modeling.

The comparative study of two different viscoelastic wall models (model v1 and model v2) in nonlinear one-dimensional simulations of blood flow are carried out [28]. Both model relate pressure and cross sectional area of vessel in different way. In the first model, pressure related to the area through instantaneous elastic response and in the model v2, pressure relates to the area through strain. The flow equations and wall viscoelasticity are solved using finite element discontinuous Galerkin method and verified using method of manufactured solution. For this the continuity and momentum equation is converted to quasi-linear form. At the inlet, the flow rate boundary conditions are applied. Initially the wave propagation study in an idealized vessel is presented. At the outlet, reflection free condition is applied and the models (two viscoelastic model and one elastic model ) are subjected to this conditions. Secondly the viscoelastic models are applied to Carotid artery. To compare the models, the inlet pressure and cross sectional area for the Carotid artery is plotted corresponds to 5th and 9th cardiac cycle. The pressure differences between models are found to be very less. The final study was done in an abdominal aorta under rest and exercise conditions and compared the results of models. The pressure pulse formed, shows very small difference in the models. The hysterisis loop for v2 showed higher value than v1. It mainly explained about the dissipative nature of v2 model.

The experimental study is conducted to examine the pulsatile blood flow through the small arteries using glycerol solution with small amount of xanthan [29]. Blood is considered as non-Newtonian fluid with very low Reynolds number. To obtain the pulsatile nature in experiment, syringe is used. The velocity through the micro channel is measured using µ-PIV method. The experiments is carried out using one parent tube and two daughter tube having equal diameter(600µm). The velocity distribution at three stations (station one is at the parent channel, station two be at entrance of bifurcation and station three is at one diameter downstream of station two) are plotted in both Newtonian and non-Newtonian fluid. At station one, both Newtonian model and non-Newtonian velocity profile is plotted and the velocity profile of latter one is found to be flatter than first one. The velocity profile of Newtonian fluid, at station two showed a shifting towards the left due to the secondary flow development in curved section. But at station three the Newtonian fluid profile is again similar to that off station one profile .The reason behind this phenomenon is that the secondary flows are decayed before the first diameter downstream when Reynolds number is less than 30. From the velocity distribution graph, the wall shear stress is calculated by substituting the shear rate value in Herschel Bulkley model. The wall shear stress values are low in small arteries compared to larger one. The results showed that in small arteries blood act as non Newtonian fluid.

For simulating the hemodynamics changes in cardiovascular system, a numerical comparison has been attempted with three different ventricular assist devices [30]. Initially, numerical model of normal cardiovascular system is developed including the heart chambers, pulmonary and systemic circulation. The simulation result is similar to the normal cardiovascular system. In the heart failure condition ventricle become weak and cannot be able to pump the required amount of blood. In this study, the diseased heart is simulated by reducing the maximum elastance in the left ventricle value to very low value (0.5mm Hg/ml). In second model, ventricular assist devices are installed to the diseased heart system. The first model is then modified using ventricular assist devices system which is connected either in parallel or in series with the cardiovascular system, individually. Three types of Ventricular assist devices are studied. They are impeller pump, displacement type and reciprocating valve pump. The impeller pump and displacement pump are installed in parallel and reciprocating valve pump in series in connection with the heart. In this study Ventricular assist devices are operated in three different control modes and developed, six cases
of simulation. The results of left arterial pressure, left ventricular pressure, left ventricular volume and wall shear stress in the six simulations are compared. In pressure comparison study the displacement pump in counter-pulsation showed lowest pressure and the displacement study in co-pulsation showed highest pressure in left ventricle. Due to this reason, displacement pump is better than impeller. In volume comparison study, displacement pump in counter-pulsation shows very less value. But in the power requirement, the impeller needs less power than displacement pump.

Toloui et al., [31] investigated the wall deformability and blood rheological properties in the wall shear stress. In this study fluid is modelled as Newtonian and non-Newtonian fluid (Carreau- Yasuda model). The wall is considered as rigid and deformable. The fluid flow is explained by Navier-Stoke equation. The fluid structure interaction is modelled using ANSYS-CFX software. Both steady state and pulsatile state is modelled. The boundary conditions like velocity inlet, no slip condition at wall and pressure zero condition at outlet are applied. In pulsatile condition time varying pressure condition at inlet and velocity profile at outlet is given. At the interface, the stress continuity and velocity continuity conditions are applied. The following conclusions are derived in steady state with rigid wall and deformable wall. It is found that the Newtonian model is not suitable in Carotid blood flow due to the following reasons. In Newtonian model, the viscosity values are lower than the apparent viscosity and also the area of back flow zone is more. Therefore the non-Newtonian model is more suitable for blood flow. Newtonian model, underestimate the wall shear stress in the interface. In pulsatile study, the blood is modelled as Carreau-Yasuda model. The dynamic viscosity is more in the middle of the flow. The wall shear stress distribution of pulsatile blood flow in both rigid and deformable models shows same pattern. The wall shear stress values of both models show significant difference at the end of two cardiac cycles. The wall shear stress showed negative values, which indicate the presence of back flow.

2.3. Fluid structure interaction

Arterial wall mechanics plays an important role in hemodynamic problem. The mechanical interaction between blood and arterial wall is mainly responsible for the propagation of pressure wave from the heart to whole body. This interaction is the main factor which help to regulate blood pressure in the body. An efficient and accurate numerical simulation of this mechanics has an important role in atherosclerosis detection and stent bypass.

A two layer structural model has been developed for artery study by taking the media and adventitia layers as a fiber reinforced composite layer [32]. They concluded that the model can be used to study arterial wall mechanics. The finite element models are applied to study fluid interaction with the operation of aortic valve [33]. The blood is considered as Newtonian fluid. In this study both left ventricle and aortic valve is modelled. The fluid structure interaction of left ventricle is simulated and with the help of these data aortic valve is modelled. From the result, it is found out that pressure difference across the valve leaflet is uniform except for a short period just before it is fully open. The fluid structure interaction of a one ruptured aneurysm, two unruptured aneurysm and without aneurysm is modelled in middle cerebral artery [34]. The blood flow in arteries is treated as laminar one. The blood is assumed as an incompressible fluid. The flow is described based on the Navier-Stoke equation. The arterial wall mechanics is explained with the help of force equilibrium equations. The arterial wall structure is modelled as a linearly elastic material with finite strain. From the result, it is found out that the shape of aneurysm is an important factor in fluid structure interaction. The periodic unsteady incompressible flow through a compliant vessel is developed by adding a time periodic function in a weakly coupled fluid interaction model [35]. The time periodic method is applied to straight, curved and bifurcating arteries. Pressure distribution of next time periods are found out from the pressure values of the previous time step. The time periodic coupling method has been compared to the decoupled and weakly coupled approaches. It is found that time periodic, coupled model gives improved computational stability compared to the weakly coupled approach. Two-way coupled FSI model is developed for the prediction of transient flow in a complaint tube [36]. The FSI model is validated by using experimental method and analytical method. Flow parameters such as pressure, axial stress, wave speed, fluid velocity etc are also studied. From the comparison study it is found that numerical model developed for complaint tube is reliable.

The arterial wall is modelled as three dimensional non linear hyperelastic material by considering blood as a shear thinning fluid. Both fluid and solid equations are combined together to form FSI. It is also observed that the selection of models for blood flow analysis is an important factor [37]. For predicting hemodynamics in large
arteries, algorithms are developed based on the finite element concept [38]. The pulse wave velocity is also calculated in artery system. The proper outlet and inlet conditions are applied in the model. At the outlet, the flux conditions are applied which is obtained from the one dimensional simulation. The FSI and CFD simulation of rigid wall results are compared. From the results it is found that the magnitude of wall shear stress is overestimated in rigid wall modeling. Another model for fluid structure interaction has been developed by considering the arterial wall as linearly viscoelastic and a cylindrical shell [39]. Fluid is considered as incompressible, viscous fluid. The fluid and structure are fully coupled using kinematic and dynamic coupling conditions. The method give good result and also it requires low computational costs. Gilmanov et al., [40] presented the numerical methods for simulating the fluid structure interaction between thin flexible tube and an incompressible fluid. The curvilinear immersed boundary method and finite element method are coupled for fluid structure interaction study. The coupled curvilinear immersed boundary method and finite element method is validated by applying in a two fluid structure interaction problems involving flexible pipes and compared the results with the previous numerical simulation and experimental results. From the comparison study it is found that the method gives good results.

3. Conclusions and future work

Based on the literature review, the following conclusions are derived.

- The consideration of a non-Newtonian behaviour of blood in small arteries will give more relevant results.
- Prediction and understanding of pressure waves have significant importance in medical diagnosis area.

From the reviews, it is also found that interaction between vessel wall and the blood will give more realistic results. However, consideration of full three dimensional FSI problem increases the computational work. Hence, one of the complexity to be faced is the change in the properties of blood and vessel. In addition, in most of the studies, the data used for the simulation are collected from the medical scan reports. Hence, collection of accurate data for the development of the model is also a big challenge.

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