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Computer modelling of electro-osmotically augmented three-layered microvascular peristaltic blood flow

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ABSTRACT

A theoretical study is presented here for the electro-osmosis modulated peristaltic three-layered capillary flow of viscous fluids with different viscosities in the layers. The layers considered here are the core layer, the intermediate layer and the peripheral layer. The analysis has been carried out under a number of physical restrictions viz. Debye-Hückel linearization (i.e. wall zeta potential ≤25 mV) is assumed sufficiently small, thin electric double layer limit (i.e. the peripheral layer is much thicker than the electric double layer thickness), low Reynolds number and large wavelength approximations. A non-dimensional analysis is used to linearize the boundary value problem. Fluid-fluid interfaces, peristaltic pumping characteristics, and trapping phenomenon are simulated. Present study also evaluates the responses of interface, pressure rise, time-averaged volume flow rate, maximum pressure rise, and the influence of Helmholtz-Smoluchowski velocity on the mechanical efficiency (with two different cases of the viscosity of fluids between the intermediate and the peripheral layer). Trapping phenomenon along with bolus dynamics evolution with thin EDL effects are analyzed. The findings of this study may ultimately be useful to control the microvascular flow during the fractionation of blood into plasma (in the peripheral layer), buffy coat (intermediate layer) and erythrocytes (core layer). This work may also contributes in electrophoresis, hematology, electrohydrodynamic therapy and, design and development of biomimetic electro-osmotic pumps.

1. Introduction

Blood circulation is an important mechanism in animal and human physiology. It is a prime requirement for sustaining life. Blood transport in cardiovascular system is an extremely sophisticated geometrical and hydrodynamic phenomenon. In the last few decades, many theoretical (analytical and computational) and experimental investigations on blood circulation have been reported. These hemodynamic studies have observed that blood flow rate typically depends on pressure difference, viscosity, and geometry of blood vessels. For example, Peskin (1977) conducted a seminal computational study of blood flow where a fast Laplace-solver was used to solve Navier-Stokes equations for the moving immersed boundaries interacting with fluid (add result of this study). Nevertheless, hemodynamics has provided a very wide spectrum for mathematical and numerical simulations. It is also an issue of great clinical interest to be addressed since many losses to life have been reported in developing countries due to abnormal blood flow and hematological disorders. Along the same line, Ku (1997) provided an excellent review on a series of complex phenomenon in the cardiovascular flows such as unsteadiness, pulsatile dynamics, turbulence, multiple branching, vessel deformability, blockages and curvature effects. Taylor et al. (1998) conducted a finite element study on blood flow. This study have highlighted the advantages of computational blood flow analysis and simulations in assisting clinicians to diagnose and address a variety of clinical vascular problems.

Blood flow in microvascular system i.e. blood microcirculation occur via venules, arterioles and capillaries. A series of studies by Pries et al. (1990, 1996) have provided important information on biophysical aspects of blood flow. It is well known that blood is a bio-rheological, incompressible, multi-component viscous fluid; and viscosity has a major role in the blood flow through circulatory system. This introduces a possibility of blood fractionation during blood circulation in three layers as blood plasma, white blood cells (WBCs) and red blood cells (RBCs) due to differences in their density and viscosity. This variation in viscosity may be due to uneven temperature distribution or composition. Viscosity has also been reported as a function of pressure and hence it influences induced shear stress. In addition, optimized tissue perfusion is only achieved when rheological properties of blood attain a specific range. Two and three-layered blood flows are an important area of investigation to present more realistic hemodynamic simulations. A three-layered Couette flow model has been presented by Chaturani and Biswas (1983) using Stokes couple stress (polar) model, where expressions were derived for estimation of linear velocity,

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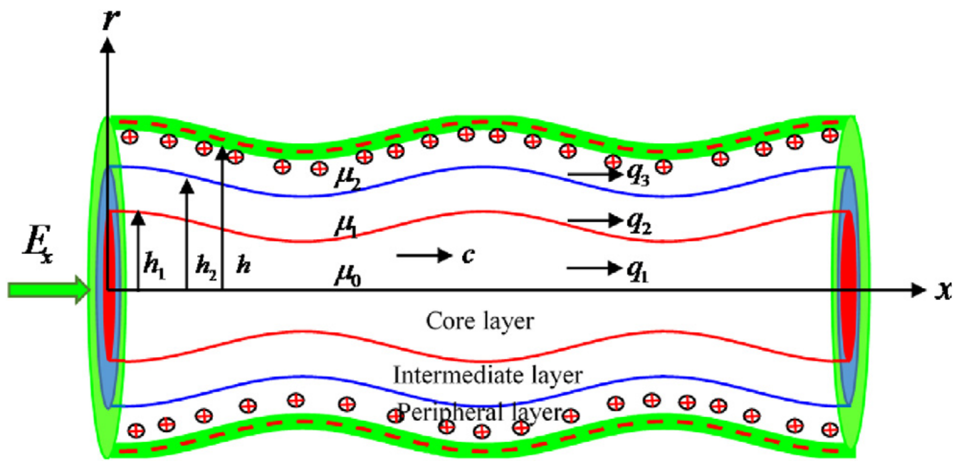

Fig. 1. A geometrical model of three layered blood flow through capillary in presence of electric field and EDL formation.

total angular velocity and effective viscosity of blood. Pralhad and Schultz (1987) developed a two layered blood flow model for stenosed arteries, simulating the core as a polar fluid and the plasma layer as a Newtonian fluid. They considered a variety of diseases including polycythemia and plasma cell dyscrasias and the influence of size effects (particle size to tube diameter) on flow dynamics has been described in detail.

Most of the transport phenomenon in biology are achieved with mechanism of *peristalsis* e.g. blood flows, phloem translocation in botany and digestive pumping. Peristalsis involves the efficient and systematic propulsion of materials in deformable vessels via rhythmic expansion and contraction of the vessel walls. As a moving boundary value problem, peristaltic hydrodynamics has attracted significant attention in engineering sciences. Brasseur et al. (1987) considered the influence of peripheral features on peristaltic pumping focusing on the coating in physiological flows. They showed that peristaltic pumping efficiency improves when the peripheral layer viscosity exceeds the inner core viscosity. However, lower efficiency, reduced trapping, and refluxes were shown with a lower peripheral layer viscosity for fixed total volume flow rate. In the literature, peristaltic two-layered and other hemodynamic two-layered flows have been discussed for a variety of fluid types. These include Casson viscoplastic fluids in stenosed vessels (Srivastava and Saxena, 1994), immiscible Newtonian fluids (Usha and Rao, 1997), Casson and Newtonian fluids (Haldar and Andersson, 1996; Srivastava and Saxena, 1995), two-phase models (Srivastava, 1996), Ostwald-de Waele power-law models, Newtonian fluid-saturated porous media models (Mishra and Rao, 2005), multiple viscosity Newtonian models (Ponalagusamy, 2007), micro-continuum models (Ikbal et al., 2009), models simulating a suspension of erythrocytes in plasma as a particle–fluid suspension and a peripheral layer of plasma as a Newtonian fluid (Srivastava et al., 2010) are few examples. Other studies on two-layered blood flows have considered heat transfer (Rattanadecho and Keangin, 2013) and/or mass transfer (Farooq et al., 2014) to elaborate the heat-conducting and diffusive properties of blood. Thus, there are very few studies in the literature which have implemented a three-layered flow model presented by Chaturani and Biswas (1983). A few efforts have been made to understand the blood flow assuming it as a three layered fluid. Elshehawey and Gharsseldien (2004) studied three-layered blood flow in deformable vessels with a stenosis, providing a more viable simulation of actual hemodynamics. Further investigations on three-layered blood flow were

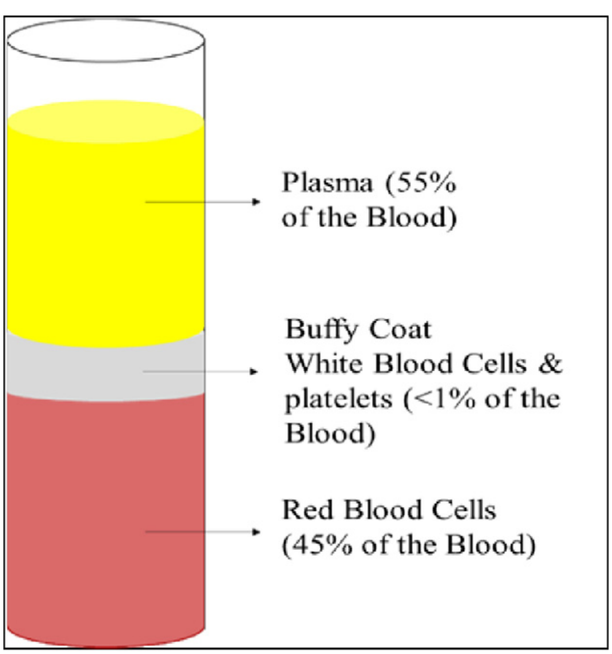

Fig. 2. Blood fractionation in three layers (Plasma in peripheral layer (yellow), WBC in intermediate layer (grey), and RBC in core layer (red)).

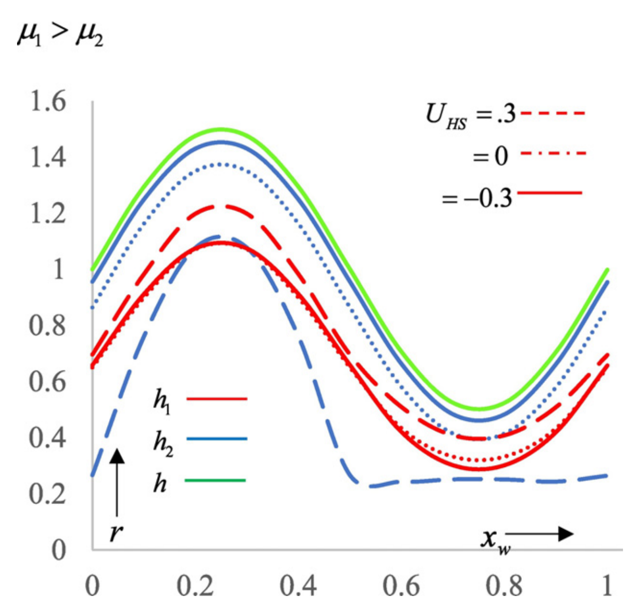

Fig. 3. The variation in interface shape h_1, h_2 corresponding to different values of Helmholtz-Smoluchowski velocities when other parameters are kept constant as$\mu_1 = 10, \mu_2 = 0.1 \overline{Q} = 0.1, \alpha = 0.65, \beta = 0.85$, and $\phi = 0.5$.