



Electrokinetically modulated peristaltic transport of power-law fluids



Prakash Goswami^{a,1}, Jeevanjyoti Chakraborty^{b,2}, Aditya Bandopadhyay^b, Suman Chakraborty^{b,c,*}

^a Department of Mathematics, Indian Institute of Technology Kharagpur, Kharagpur 721302, India

^b Advanced Technology Development Centre, Indian Institute of Technology Kharagpur, Kharagpur 721302, India

^c Department of Mechanical Engineering, Indian Institute of Technology Kharagpur, Kharagpur 721302, India

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ABSTRACT

The electrokinetically modulated peristaltic transport of power-law fluids through a narrow confinement in the form of a deformable tube is investigated. The fluid is considered to be divided into two regions – a non-Newtonian core region (described by the power-law behavior) which is surrounded by a thin wall-adhering layer of Newtonian fluid. This division mimics the occurrence of a wall-adjacent cell-free skimming layer in blood samples typically handled in microfluidic transport. The pumping characteristics and the trapping of the fluid bolus are studied by considering the effect of fluid viscosities, power-law index and electroosmosis. It is found that the zero-flow pressure rise is strongly dependent on the relative viscosity ratio of the near-wall depleted fluid and the core fluid as well as on the power-law index. The effect of electroosmosis on the pressure rise is strongly manifested at lower occlusion values, thereby indicating its importance in transport modulation for weakly peristaltic flow. It is also established that the phenomenon of trapping may be controlled on-the-fly by tuning the magnitude of the electric field: the trapping vanishes as the magnitude of the electric field is increased. Similarly, the phenomenon of reflux is shown to disappear due to the action of the applied electric field. These findings may be applied for the modulation of pumping in bio-physical environments by means of external electric fields.

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1. Introduction

Peristaltic flow is one of the most important physiological transport mechanisms in living organs and organisms (Kim and Bush, 2012; Lauga and Powers, 2008). In this mechanism, fluid is transported through narrow passages by a progressive wave generated by periodic contraction and expansion of the surrounding medium. For example, peristalsis brings about the movement of food in the alimentary canal, the flow of blood in the passages of circulatory system, and the flow of body fluids as an integral part of the natural waste-management system of the body; most importantly, peristaltic transport is a crucial step in the process leading to the procreation of life itself. The study of such peristaltic flows, albeit in model geometries, has had a long and illustrious history and new inroads continue to be made in order to better elucidate the fundamental flow patterns and fluid physics involved in these flows, such as ‘trapping’ and ‘reflux’ (Böhme and Müller, 2013; Brasseur et al., 1987; Burns and Parkes, 1967; Ceniceros and Fisher, 2012; Hung and Brown, 1976; Jaffrin and Shapiro, 1971;

Jiménez-Lozano et al., 2009; Latham, 1966; Manton, 1975; Misra and Pandey, 1999; Pozrikidis, 1987; Provost and Schwarz, 1994; Rao and Usha, 1995; Shapiro et al., 1969; Shukla et al., 1980; Takabatake et al., 1988; Takagi and Balmforth, 2011a, 2011b). The seminal work of Shapiro et al. (1969) on the peristaltic pumping of liquid, under the assumption of low Reynolds number and long wave length approximation, laid the foundation for further studies to reveal the richness of the peristaltic mechanism with increasing detail and rigor. From a physiological perspective, accounting for the rheology of the fluid (Newtonian or non-Newtonian (Böhme and Müller, 2013; Misra and Pandey, 1999)) evolves coherently within the framework laid down by Shapiro et al. (1969). It is to be noted that relatively few studies describe these phenomena beyond the low Reynolds number hydrodynamics and long wave length approximation (Li and Brasseur, 1993). Furthermore, some recent studies consider the effect of elasticity of the fluidic pathway (Takagi and Balmforth, 2011a) on fundamental peristaltic pumping. The possibility of exploiting this elegant transport mechanism in engineering systems has led researchers to explore the applicability of peristaltic pumping in the realm of microfluidic devices (Aboelkassem and Staples, 2013, 2012; Chu, 2002; Husband et al., 2004; Moroney et al., 1991; Nguyen et al., 2000; Selverov and Stone, 2001; Tripathi and Bég, 2014).

In spite of the tremendous progress made into the straightforward modeling of peristaltic flows, relatively little interest has been devoted to the possibilities of augmenting such peristaltic flows through

* Corresponding author at: Advanced Technology Development Centre, Indian Institute of Technology Kharagpur, Kharagpur 721302, India.

E-mail address: suman@mech.iitkgp.ernet.in (S. Chakraborty).

¹ Current affiliation: Department of Mechanical Engineering, Indian Institute of Technology Kharagpur, Kharagpur 721302, India.

² Current affiliation: School of Mathematics, University of Birmingham, Birmingham B15 2TT, United Kingdom.

extraneous measures. One such example is the augmentation of peristaltic transport by electroosmotic means (Chakraborty, 2006). That study, however, is restricted to Newtonian fluids, and, as such, is of limited applicability in the context of bio-fluids. Regarding this, the study of electrokinetic flow of non-Newtonian fluids are mostly involved in pumping through straight microchannels and tubes (Afonso et al., 2009; Das and Chakraborty, 2006; Zhao and Yang, 2013a, 2013b, 2011; Zhao et al., 2013). The aim of the present work, accordingly, is to investigate the effect of electroosmosis on the peristaltic flow of non-Newtonian fluids. For our study to be immediately applicable to the particular case of blood flow, we add an extra level of sophistication to our model. We first recognize that blood has very low apparent viscosity in small diameter tubes (Fahraeus and Lindqvist, 1931; Haynes, 1960). The viscosity increases with the increase in the tube diameter and approaches an asymptotic value after the diameter reaches a certain maximum (~0.5 mm). This is the Fahraeus–Lindqvist effect which we incorporate into our model by invoking the marginal zone theory. In simple terms, we represent the cell-depleted peripheral region of the blood, predominantly consisting of plasma, as a Newtonian fluid, and the undepleted core region by a power-law model, with appropriate boundary conditions imposed to ensure physical continuity across these regions. It is important to mention here that although in principle more complex (and, possibly, more realistic) constitutive models could be employed to mimic the fluid rheology, here we employ the power-law model for its inherent simplicity in capturing the essential physics of concern.

Before proceeding into a detailed discussion of the modeling aspects, it is important to note that, the peristaltic motion itself is contingent on the elastic nature of the walls guiding the fluid flow. In the case of blood vessels, for example, of the four primary structural components making up the vessel walls: endothelial cell lining, elastic fibers, collagen fibers and smooth muscles (ordered from the inside out), it is only the smooth muscles which contribute to the elastic nature of the walls (Li, 2004). However, in the current study, we do not capture such elastic properties explicitly in our model. Rather, only the phenomenological local contraction and expansion contributing to the peristaltic fluid motion is taken into account. Nevertheless, we do take cognizance of the fact that the glycocalyx layer of the endothelial cell lining bears a negative charge which contributes towards the electrochemical properties of the surface (Liu and Yang, 2009) and which, in turn, is crucial for peristaltic flow augmentation.

Despite these underlying simplifications, the present model highlights the possibility of employing electrokinetics towards the design of embedded bio-chips for in-vivo diagnostic sensing on a real-time basis together with targeted drug delivery. Such futuristic devices are poised to be the next natural step in the bioengineering realm after devices like GlucoWatch (Tamada et al., 1995; Tierney et al., 2001) and similar ones using reverse iontophoresis (Landers, 2003; Lee, 2008; McCormick et al., 2012) which already exploit electroosmotic flow for glucose monitoring and drug delivery, albeit in strictly non-in-vivo modes.

2. Mathematical formulation

We consider a bio-fluid flowing through a cylindrical elastic tube of uniform undeformed radius r_0 under the dual influences of (i) peristaltic motion of the tube wall and (ii) electroosmotic body force as depicted in Fig. 1. There are two distinct regions in the fluid: the peripheral region which is adjacent to the tube wall and the inside core region. The walls of the tube carry a surface charge which is balanced by an equal and opposite charge in the fluid maintained through a distribution of ions: co-ions which have the same charge as that of the surface and counterions (outnumbering the co-ions) which have the opposite charge. The surface charge together with this distribution of ions is referred to as the electrical double layer (EDL). The electrokinetic body force arises as a result of the effect of the externally applied

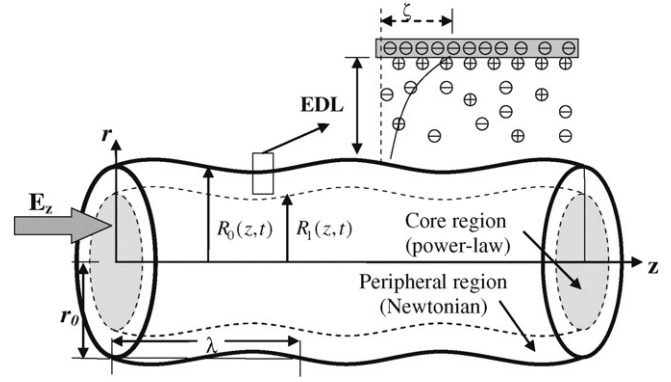


Fig. 1. Schematic depiction of the tubular passage and electrical double layer (EDL) formation close to the walls. In this particular depiction, the walls are considered to be negatively charged. Consequently, the peripheral (Newtonian fluid) and core region (Power-law fluid) have a net positive charge which balance the surface charge.

electrical field on this EDL: specifically, on that part of the ionic distribution – the diffuse layer – which is mobile.

2.1. Electro-hydrodynamic equations

The fluid transport is represented by the general Cauchy's equation of motion:

$$\rho_f \frac{D\mathbf{u}}{Dt} = -\nabla p + \nabla \cdot \boldsymbol{\tau} + \mathbf{F}_{EK}, \quad (1)$$

where ρ_f is the fluid density, $\boldsymbol{\tau}$ is the deviatoric stress tensor, p is the excess pressure, and \mathbf{F}_{EK} represents the electrokinetic body force (the exact form will be detailed later). For a consistent description in accordance with the requirements of the conservation of mass, the unknown velocity and pressure fields should, in addition, satisfy the continuity equation:

$$\frac{\partial \rho_f}{\partial t} + \nabla \cdot (\rho_f \mathbf{u}) = 0. \quad (2)$$

Besides the details of the electrokinetic body force, Eqs. (1) and (2) need to be closed by the specification of constitutive models to represent the stress in terms of the velocities consistent with the behavior of the fluid material.

As already mentioned in the introductory discussion, for the specific case of blood flow in narrow confinements, the wall-adjacent region being predominantly cell-free plasma, is characterized by Newtonian fluid behavior while the undepleted core region is characterized by non-Newtonian fluid behavior. Thus, the domain of the flow is considered as: $0 \leq r < R_1(z, t)$ (the boundaries are functions of the axial coordinate and time owing to the wave-like deformation at the surface) for the core region, $R_1(z, t) \leq r < R_0(z, t)$ for the peripheral region and $r = R_0(z, t)$ for the boundary. It is important to note that $R_i(z, t)$, ($i = 0, 1$) represent the domain limits in the deformed state in the peristaltic motion. For the sake of the current model development, we choose the Ostwald-de-Waele power-law model to represent the fluid in the core region. In mathematical terms, the constitutive equations for the fluids in the two regions are (Bird et al., 1981):

$$\boldsymbol{\tau} = \begin{cases} \tilde{\mu}_C(\dot{\boldsymbol{\gamma}})\dot{\boldsymbol{\gamma}}, & \text{Core region} \\ \mu_N \dot{\boldsymbol{\gamma}}, & \text{Peripheral region} \end{cases} \quad (3)$$

where $\tilde{\mu}_C(\dot{\boldsymbol{\gamma}}) = k_p |\dot{\boldsymbol{\gamma}}|^n$ is the apparent power-law viscosity of the non-Newtonian fluid, $\dot{\boldsymbol{\gamma}} \equiv \|\dot{\boldsymbol{\gamma}}\| = \sqrt{\boldsymbol{\dot{\gamma}} \cdot \boldsymbol{\dot{\gamma}}}$ is the norm of the strain rate tensor $\boldsymbol{\dot{\gamma}} (= \frac{1}{2} \{\nabla \mathbf{u} + (\nabla \mathbf{u})^T\})$, k_p is the consistency index, n is the fluid behavior index, and μ_N is the viscosity of fluid in the peripheral region. Note that

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