Journal of Molecular Liquids 215 (2016) 472–480

Contents lists available at ScienceDirect

Journal of Molecular Liquids

journal homepage: www.elsevier.com/locate/molliq

Electrokinetic effects on pressure driven flow of viscoelastic fluids in nanofluidic channels with Navier slip condition

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ARTICLE INFO

Article history:
Received 4 July 2015
Received in revised form 31 December 2015
Accepted 11 January 2016
Available online xxxx

Keywords:
Micro/nanofluidic
Electrokinetic
Viscoelastic fluid
EDL
Slip

ABSTRACT

The flow of viscoelastic fluids through micro/nanofluidic systems is an important issue in the biological applications. The electrokinetic effects on pressure driven flow of a viscoelastic fluid through a nanochannel with slip boundary condition are investigated in this study. The rheological behavior of the viscoelastic fluids is described using constitutive equations and the linear Navier's law is employed to consider the slip condition on the channel walls. The Debye–Hückel linearization is employed to obtain a closed form analytical solution for the velocity distribution and induced electric field as function of slip parameter, Reynolds number, viscoelastic parameter, dimensionless Debye–Hückel parameter and dimensionless zeta potential. It is found that the induced electric field increases with increasing both dimensionless zeta potential and dimensionless slip coefficient while it decreases with increasing dimensionless Debye–Hückel parameter.

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1. Introduction

Due to smaller Reynolds numbers in microfluidic systems, the flow is assumed to be laminar or simply Stokes flow. This assumption is also valid in nanochannels with/without slip velocity at the walls. In the case of liquid transport through nanochannels, the Knudsen number can be used to check the continuum assumption which is defined as the ratio of the interaction length of molecules and the system characteristic length [1–4], i.e. $Kn = 1/L$. During transportation of liquids through nanochannels, the walls of the channel become charged due to the adsorption or dissociation of ions in the region close to the walls. As a result, a liquid layer is developed with a net charge in this region, that is, electric double layer (EDL) [5,6]. The first layer has an opposite charge on the charged wall and is strongly attracted towards the wall. Whereas, the second layer is poorly attracted towards the wall and can move along the channel and is actually a mobile part on the EDL. So the ions in the mobile part of the EDL undergo a net migration in the opposite direction and carry solvent along with them, thereby causing the movement of the solvent. When the charged surface and the diffuse part of EDL move relative to each other, an electric field is created. This electric field or a pressure gradient along the channel causes the flow through the channel. These two types of flow are called as electroosmotic flow and pressure driven flow, respectively. For the pressure-driven flow, the ions in the channel flow in the direction of pressure gradient, and result in an electric current called streaming current and a potential difference along the channel called streaming potential. The streaming potential drives the ions in the channel to move in the direction opposite to the pressure-driven flow and result in a current called conduction current. So the presence of EDL leads to an electrical force opposed to the pressure driven flow.

For bio-fluids, the suitable constitutive model may be used. For fluids with viscoelastic behavior, an appropriate constitutive equations need to be considered. For instance, the PTT and FENE-P models, are typically used for modeling fluids such as blood [7,8], saliva, synovial fluid [9] or other biofluids containing long chain molecules. During the past two decades, micro/nanofluidic systems have attracted a large amount of attention due to their vital applications in engineering, biology and medicine. The microfluidic and nanofluidic imply fluid motion through or past structures with a size smaller than 100 μm and 100 nm, respectively. The first motivation in studying micro/nanofluidic systems is to move polymers used in DNA separation through solid devices. The fluid flow inside micro/nanofluidic channels was found different from the flow in typical macrosclae channels. The interesting phenomena are to be expected as fluidic dimensions go down the characteristic dimensions. These phenomena, which are prominent in micro/nanofluidic channels, arise from the interactions between working fluid and the walls that include surface tension effects, boundary slip, and electric charge interactions. Since the number of molecules becomes fewer in nanofluidic devices, transport in such devices differs from the microfluidic ones. Electrokinetic transport differs from microfluidic systems due to the influence of the electrical double layer. Wei et al. [10] investigated the purely electroosmotic flow in a

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http://dx.doi.org/10.1016/j.molliq.2016.01.034
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slit microchannel and concluded that the calculated electroosmotic values basically agree with the experimental ones. The electroosmotic and pressure driven flows for power-law biofluids in microfluidic channels were studied by other researchers [11,12]. Abgrall and Nguyen [13] presented the current state of the art of nanofluidic fabrication and discussed the key transport phenomema in nanochannels. Perry and Kandlikar [14] reviewed the most commonly used methods for nanostructuring and nanochannel fabrication used in nanofluidics. Schoch et al. [15] focused on the physical properties and operational mechanisms of the most common structures like nanometer-sized openings and nanowires in solution on a chip. They described physical mechanisms on the nanometer scale where new phenomena occur. Yuan et al. [16] reviewed theoretical and experimental studies related to the electrokinetic transport in small fluidic nanochannels. They focused on the effect of the electric double layer on the EOF, electric current, and electrophoresis of charged analytes. Pennathur and Santiago [17] reported analytical and numerical models for nanochannel electrophoretic transport and separation of neutral and charged analytes. They obtained analytical expressions for electroosmotic flow, species transport velocity, streamwise–transverse concentration field distribution, and ratio of apparent electrophoretic mobility for a nanochannel to (standard) ion mobility.

In the literature it is repeatedly stated that for Kn < 0.01 the continuum model for the flow is applicable and for 0.01 < Kn < 0.1, the slip boundary condition has to be considered to use continuum model for the flow [3]. Since nanofluidic channels might deal with Knudsen numbers smaller than 0.01 the slip boundary condition is implemented here.

One of the known approaches for modeling the boundary slip was introduced by Navier [18]. In the Navier's approach there is a relative motion between fluid flow and solid wall, and the tangential velocity of flow is proportional to the velocity gradient in the direction perpendicular to the solid wall. In micro/nanofluidics the EDL is usually very small compared with the channel height and in such a case the bulk flow can be modeled accurately considering the linear Navier slip boundary condition at the wall.

Chatzimina et al. [19] showed that in an annular channel for pure pressure-driven flows of viscoelastic fluids, there is a nonlinear slip on the wall as $u_w = \alpha {\tau_w}^m$, in which, values of m range from 1 (Newtonian) to 4. However, for modeling the boundary slip in the slit micro/nanofluidics the literature is scarce and we could not find any experimental data regarding the relationship between velocity and velocity gradient on the walls and hence the Navier slip condition is used in the present study.

Yang and Kwok [20] investigated the simultaneous influence EDL thickness and Navier's slip condition on the flow in circular microchannel using Poisson–Boltzmann distribution, which refers to the relationship between ionic concentration and potential in an infinitely aqueous phase. The effect of slip boundary condition on pressure driven flow in micro/nanofluidics was studied by Jing and Bhushan [21]. They employed the Navier's formulation for modeling the boundary slip. Afonso et al. [22] analytically examined the effect of boundary slip on the combined electroosmotic and pressure driven flow in a microchannel for a viscoelastic fluid. Their results demonstrated that both the presence of the boundary slip and viscoelasticity induce an increase in the flow rate. As of late, Ferrás et al. [23] investigated the effect of boundary slip on the electroosmotic and pressure driven flow in the annular space between two cylinders.

In this study, the effect of boundary slip on the purely pressure driven flow of a viscoelastic biofluid through a nanofluidic channel is investigated. The rheological behavior of the fluid is described by viscoelastic constitutive equations which are related to the PTT model. Since the width of the channel is much larger than the height of the rectangular shaped channel, it may be presented by a two dimensional slit. Flow is assumed to be fully developed and a linear slip is modeled on the walls using the Navier's law. Assuming no overlapped EDL, the potential distribution inside the channel is obtained using Poisson–Boltzmann model and the results are employed in the Navier–Stokes momentum equation to give the induced electric field and velocity distributions.

2. Problem formulation

Fully developed pressure-driven incompressible laminar flow of a viscoelastic fluid confined by a slit nanofluidic channel with height of H is studied. A schematic of the problem along with the coordinate system is depicted in Fig. 1. In order to study the effect of boundary slip, a linear slip using the Navier's law is considered on the liquid–wall interface. It is assumed that the thermophysical properties are constant which require the temperature variations to be less than 10 K. Also it is assumed that the walls are negatively charged and the liquid contains an ideal solution of fully dissociated symmetric salt, the EDLs formed on the walls do not overlap, and the temperature variation over the channel cross section is negligible compared with the absolute temperature. This assumption is completely in agreement with the assumption of constant thermophysical properties because a temperature variation of less than 10 K is indeed negligible compared with the typical absolute temperatures of more than 300 K. Using this assumption, the potential field and the charge density may be calculated on the basis of an average temperature. The middle part of the infinitely long channel, that is, the part far away from the source of liquid, is studied, so the Donnan potential is not considered. Also the effect of surface charge on the electric conductivity and the slip length is assumed to be negligible.

2.1. Electrical potential and ionic concentration distribution

The relationship between the electric potential and the volumetric net charge density in the EDL can be expressed using one dimensional Poisson equation

$$\frac{d^2\psi}{dy^2} = -\frac{\rho_e}{\varepsilon} \tag{1}$$

where ψ is the electric potential due to the EDL at the equilibrium condition corresponding to no applied external field, ε is the permittivity of the electrolyte and $\rho_e$is the volumetric net charge density that can be expressed as

$$\rho_e = ze(n_+ - n_-) \tag{2}$$

where z is the valence number of ions, e is the proton charge, and n_+ and n_- being the ionic number concentration of counter-ion and co-ion, respectively. For describing the relation between ionic number concentration and electrical potential in the EDL the thermodynamic equilibrium is used as

$$\frac{1}{n_\pm}\frac{dn_\pm}{dy} = \mp\frac{ze}{k_B T}\frac{d\psi}{dy} \tag{3}$$

with $k_B$and T_{av}being the Boltzmann constant and the average temperature respectively. Noting that the ionic number concentration in the middle of the channel for no overlapped EDL is equal to the ionic number concentration for the neutral electrolyte, n_0, and the electric potential caused by EDL is zero, $\psi = 0$, by integrating Eq. (3) the Boltzmann equation is derived as follows

$$n_\pm = n_0\, e^{\mp\frac{ze}{k_B T_{av}}\psi}. \tag{4}$$

Substituting Eqs. (2) and (4) into Eq. (1) the Poisson equation can be rewritten as

$$\frac{d^2\psi}{dy^2} = \frac{2n_0 ez}{\varepsilon}\sinh\left(\frac{ez\psi}{k_B T_{av}}\right). \tag{5}$$

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