Colloids and Surfaces A: Physicochem. Eng. Aspects 513 (2017) 57–65

Contents lists available at ScienceDirect

Colloids and Surfaces A: Physicochemical and Engineering Aspects

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

Pulsatile electroosmotic flow in a microcapillary with the slip boundary condition

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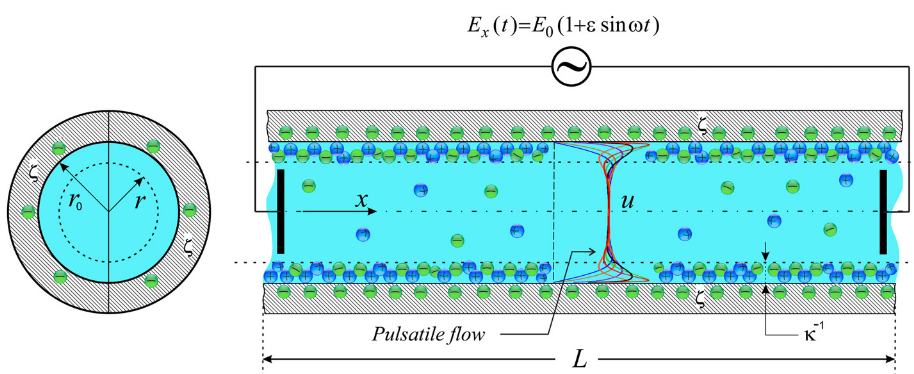
HIGHLIGHTS

- The pulsatile electroosmotic flow (PEOF) in a circular microchannel is analyzed.
- The slip boundary condition is taken into account.
- Low and high zeta potentials are assumed in the analysis.
- Slippage at the wall increases notably the volumetric flow rate.

GRAPHICAL ABSTRACT



ARTICLE INFO

Article history:
Received 8 July 2016
Received in revised form
20 September 2016
Accepted 28 October 2016
Available online 1 November 2016

Keywords:
Transient electroosmotic flow
Microcapillary
Slip boundary condition
Pulsatile electroosmotic flow
High zeta potential

ABSTRACT

The pulsatile electroosmotic flow (PEOF) of a Newtonian fluid in a circular microchannel with slippage at the surface is theoretically analyzed. It is assumed that the electroosmotic flow is started from rest by the sudden imposition of a time-dependent external electric field. The PEOF is controlled by the following dimensionless parameters: the normalized slip length, δ, defined as the ratio of the Navier length to the microcapillary radius; the angular Reynolds number, R_ω, which quantifies the competition between the time scale for the diffusion of momentum across the microcapillary and the externally imposed time scale due to the oscillatory electric field; the parameter ε, which characterizes the amplitude of the oscillating electric field; the electrokinetic parameter $\bar{\kappa}$, defined as the ratio of the characteristic length scale to the Debye length; and the ionic energy parameter, α, which compares the electric potential at the surface to the thermal potential. The analysis is conducted for low and high zeta potentials; in the former case, an analytical solution is obtained, whereas in the latter, a numerical solution is found. The results reveal that the volumetric flow rate is notably increased in a microchannel with slippage at the wall surface.

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http://dx.doi.org/10.1016/j.colsurfa.2016.10.064

1. Introduction

The transport of small volumes of fluid in microchannels has rapidly become a research field of fundamental interest because of its applications in medical science, biology, and many other areas [1]. In recent years, electroosmosis has become well established as a micropumping technique used in many devices, such as the well-known Lab-On-a-Chip (LOC), intravenous drug delivery systems, and biochemical reactive platforms [2]. In this context, electroosmosis is widely used for manipulating and controlling fluid flows in channels with lengths of less than a millimeter and is achieved by means of the electrostatic interaction between an external electric field and an electric double layer (EDL). An EDL is created when an electrolyte comes in contact with a dielectric material. This interaction generates an electric force near the wall, provoking fluid motion, which is subsequently transmitted to the bulk fluid by viscous forces [3].

Electroosmotic pumping is an important mechanism for the transport and control of flows. Typically, the key parameters that determine the pumping performance are (i) the magnitude of the electric field that is applied externally, (ii) the cross-sectional dimensions of the channel, (iii) the surface charge density of the microchannel surface and (iv) the ion density and PH of the working fluid [4].

One method of augmenting the volumetric flow rate in a microchannel is by increasing the magnitude of the applied electric field; however, this can cause an increase in the temperature of the fluid as a result of the Joule heating effect, which is undesirable. Therefore, other mechanisms must be used to achieve higher volumetric flow rates. To this end, another technique for controlling fluid flow is to use pulsating forces. Although in the specialized literature, flow enhancement in pulsatile flows has been studied at the macroscale [5,6], this concept has only recently been studied in the context of electroosmotic flow by Chakraborty and colleagues [7,8], who conducted a study of pulsatile electroosmotic flows to characterize and achieve control of the periodic characteristics of the mass flow rates by employing pulsating electric fields in microchannels. Their results highlight the importance of using pulsating forces to control mass flow rates and provide important guidance for the design of practical microdevices. To the best of the authors' knowledge, the theoretical works discussed above are the first to study electroosmotic flow under pulsating electric fields; in these works, low zeta potentials and the no-slip condition at the microchannel wall were considered.

Additionally, for the achievement of higher volumetric flow rates in such applications, the use of hydrophobic materials for the fabrication of microchannels has become common [9]. In this context, hydrophobic surfaces also exhibit electrokinetic phenomena [10,11], which are very relevant to the analysis of optimal designs for microchannels to be used in microfluidics. Thus, considerable research on electroosmotic flows (EOFs) under steady-state conditions has been conducted under the assumption of the slip boundary condition at the microchannel wall [12–19]. In particular, the oscillating electroosmotic flow for cylindrical and Cartesian coordinates, with the Navier slip condition [20] was studied by Jun and Kwok [21,22], under the Debye–Hückel approximation, where periodic solutions for the flow field were obtained.

In this work, we develop a theoretical analysis for determining the flow dynamics of a PEOF by considering the slip condition and the application of an external pulsating electric field. To generate the PEOF, we assume that the externally applied electric field, $E_x(t)$, depends on time t and can be described as the superposition of a steady electric field E_0 and a zero-mean electric field noise component, $\varepsilon E_0 f(t)$:

$$E_x(t) = E_0\left[1 + \varepsilon f(t)\right], \tag{1}$$

where ε is a dimensionless constant that determines the amplitude of the electric field fluctuations and it is assumed that $f(t) = \sin(\omega t)$, where ω represents the angular frequency.

In addition, the Navier slip boundary condition at the interface between the fluid and the microchannel surface is considered, and the slip velocity $\mathbf{u}_s$ obeys the following relationship [23,24]:

$$\mathbf{u}_s = (\mathbf{u}\cdot\mathbf{n})\mathbf{n} + \lambda_N\{\bar{\bar{\gamma}}\cdot\mathbf{n} - \left[(\bar{\bar{\gamma}}\cdot\mathbf{n})\cdot\mathbf{n}\right]\mathbf{n}\},$$

where $\mathbf{u}$ is the velocity field, λ_N is the Navier slip length, $\mathbf{n}$ is the unit vector normal to the microchannel surface pointing in toward the fluid, and $\bar{\bar{\gamma}}$ is the rate of strain tensor given by $\bar{\bar{\gamma}} = \nabla\mathbf{u} + (\nabla\mathbf{u})^T$. Due to the impermeability boundary condition, the velocity component normal to the surface of the microchannel is zero and consequently $(\mathbf{u}\cdot\mathbf{n}) = 0$, and the above statement of the Navier slip boundary condition can be written as,

$$\mathbf{u}_s = \lambda_N\{\bar{\bar{\gamma}}\cdot\mathbf{n} - \left[(\bar{\bar{\gamma}}\cdot\mathbf{n})\cdot\mathbf{n}\right]\mathbf{n}\}, \tag{2}$$

To our knowledge, in the specialized literature on EOFs, no previous investigation of the PEOF, with slippage at the microcapillary wall has been conducted. In contrast to the aforementioned researches, in this work we analyze the transient and periodic solutions of the PEOF, for high and low zeta potentials at the surface of the microchannel wall, showing the dynamic behavior of the flow. For high zeta potentials, the solution is numerically determined, while that for low zeta potentials, we find an analytical solution, from which we determine asymptotic results that elucidate the dynamic behavior of the PEOF as the times goes on. In this work, we show that the flow can be substantially increased by using hydrophobic surfaces and that the frequency of the external pulsatile electric field determines the response of the flow field, in analogy with pulsatile flows induced by oscillating pressure gradients, which have been widely studied in macroscale devices.

2. Theoretical modeling

2.1. Physical model

Fig. 1 shows the physical model analyzed in this work. Let us consider a symmetric $(z:z)$ electrolyte in a circular microchannel with a hydrophobic surface whose zeta potential, ζ, is uniform. The length of the microchannel, L, is much greater that its radius, r_0. The flow is driven by a pulsatile electroosmotic force induced by the simultaneous effect of the EDL formed at the interface between the liquid and the microchannel surface and the sudden imposition of an external time-dependent electric field given by Eq. (1). A 2D cylindrical coordinate system (r, x) is adopted, in which the origin is placed at the left end of the capillary. Additionally, we make the following assumptions: (i) The Debye length, denoted by λ_D or κ^{-1} and defined as $\lambda_D = \left(\epsilon k_B T/2e^2z^2n_\infty\right)^{1/2}$ [25], is much smaller than r_0; here, ϵ, k_B, T, e, z and n_∞ denote the dielectric permittivity of the solvent, the Boltzmann constant, the absolute temperature, the elementary charge, and the valence and bulk concentrations, respectively. (ii) The net charge density in the EDL follows the well-known Boltzmann distribution, which remains valid if the frequency of the external electric field is not very high (e.g., less than 1 MHz) [26].

2.2. Governing equations

2.2.1. Electric potential

According to Hsu et al. [27], the characteristic time scale of the electro-migration in the EDL is on the order of 10^{-8}–10^{-7} s, whereas the characteristic time scale associated with the evolution of the electroosmotic flow is on the order of 10^{-5}–10^{-3} s [28]. Hence, this latter time scale is at least two orders of magnitude greater than the