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Control of flow rate and concentration in microchannel branches by induced-charge electrokinetic flow

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ABSTRACT

This paper presents a numerical study of controlling the flow rate and the concentration in a microchannel network by utilizing induced-charge electrokinetic flow (ICEKF). ICEKF over an electrically conducting surface in a microchannel will generate vortices, which can be used to adjust the flow rates and the concentrations in different microchannel branches. The flow field and concentration field were studied under different applied electric fields and with different sizes of the conducting surfaces. The results show that, by using appropriate size of the conducting surfaces in appropriate locations, the microfluidic system can generate not only streams of the same flow rate or linearly decreased flow rates in different channels, but also different, uniform concentrations within a short mixing length quickly.

1. Introduction

Concentration gradients of biological or chemical solutions are most useful in chemotaxis studies of cells and biomolecules which migrate in solutions by sensing the surrounding concentration [1]. Practically, it is important to generate concentration gradients of any required profile and maintain the gradients spatially and temporally constant. Several technologies based on microchip have been developed to mix different solutions. Serial dilution devices are widely used to generate concentration gradients in stepwise microfluidic networks [1,2]. The first microfluidic network of this kind was created by Dertinger's group [1]. In this network, two or more solutions were divided into capillaries in a multi-step network to generate streams carrying different concentrations. Various concentration profiles can be generated by adjusting the flow rates of the source streams or the length and width of the capillaries. Efforts have been made to generate more shapes of the concentration gradients and compact the networks [3–6] because diffusion in laminar flow needs long channels. However, the structures of such serial dilution networks are still complex for fabrication and it takes long time (3–10 min) to obtain the desired gradients.

Other alternative microfluidic-based approaches [7,8] also allow mixing of different solutions, including using passive pump and evaporation [9], flow resistance [10,11] or combination of chips [12]. These techniques are able to generate controllable concentration gradients; however, they suffer from either time consumption or complicated structures. Electrokinetic flow [13,14] in microchannels can be realized by applying an electric field along the channel. It has simple structure and is easy to control. In electrokinetic flow, the mixing of different solutions is achieved by switching electroosmotic flow of the source streams in T-shaped or Y-shaped microchannels [15–17] or by applying a high voltage to introduce flow instability [16]. However, switching and controlling the electric fields in a microchannel network may be complicated, and high electric field is not appropriate for biological solutions.

Induced-charge electrokinetic flow generated by an electrically conducting object immersed in an aqueous solution provides another way of solution mixing [18–21]. When in contact with an aqueous solution, a solid surface usually carries electrostatic charges, and the charged surface will attracts counterions in the liquid to balance the charges at the solid surface. The charged surface and the layer of liquid containing the balancing charges are called the electric double layer (EDL). When an electrically conducting object is placed in an electric field, opposite charges are induced on different parts of the conducting surface which will in turn attract the counterions in the liquid to form dipolar electrical double layers around the conducting surface. The interaction between the applied electric field and the net charges in the dipolar EDL will generate vortexes near the conducting object [18]. The induced-charge electrokinetic flow circulations are very strong and can greatly enhance the flow mixing. [21–23]. In this study, we wish to develop a simple method that uses the induced-charge electrokinetic flow to produce different, uniform concentrations in different microchannel

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branches within a short time and a short distance. In addition, the vortexes produced by the ICEKF will be used to control the flow rates in different microchannel branches.

2. Physical and mathematical models

The mixing system in this study, as illustrated in Fig. 1, consists of two incoming channels, a main channel and three branch channels, all have a rectangular cross-section. The thicker lines in Fig. 1 show locations where conducting surfaces are located. Two incoming channels have a width of 75 μm and a length of 600 μm; the width of the main channel is 150 μm and the length is $H = 300$ μm. The three branch channels have a width of 50 μm and a length of 600 μm and are separated by a distance $d = 250$ μm. The height of all channels is 1050 μm. The dimensions of the conducting surface in the main channel are L_1 and L_2, and the lengths of conducting surface at the inlet of the second branch channels are L_3 and L_4.

The electrical potentials at the three outlets are set to be zero, and two voltages are applied at the two inlets in order to generate and control the electroosmotic flow. A solution containing sample specie with a concentration C_0 comes from inlet 1. The solution from the inlet 2 is the sample solution except it does not contain the sample molecules, i.e., the sample concentration is zero. After mixing in the main channel, the mixed liquid flows out from three branch channels in different concentrations.

2.1. Electric field

Once the voltages are exerted at the inlets of the system, the applied potential through the system is governed by Laplace's equation [13].

$$\vec{\nabla}^2 \Psi_e = 0 \tag{1}$$

And the local applied electric field strength is

$$\vec{E} = -\vec{\nabla}\Psi_e \tag{2}$$

The exerted electric field will make the electrons move in the conducting surfaces to form an equal but opposite induced electric field to balance the exerted electric field. That is:

$$\vec{E}_i = -\vec{E} \tag{3}$$

where $\vec{E}_i$ is the induced electric field in the conducting surface. By Eq. (3), one can also have:

$$\nabla \zeta_i = -\nabla \Psi_e \tag{4}$$

where ζ_i is the induced zeta potential of the conducting surfaces.

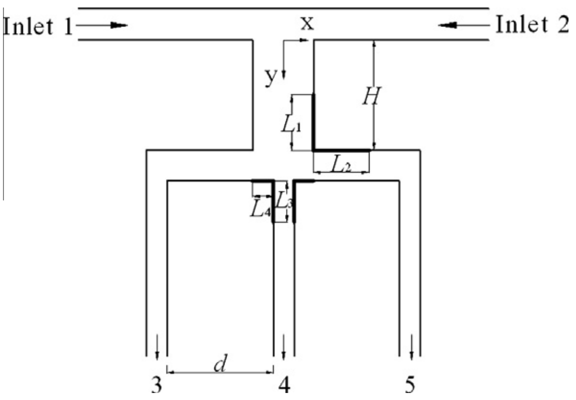

Fig. 1. Schematic of the microfluidic system with electrical conducting surfaces; thicker lines indicate the conducting surfaces.

By integrating Eq. (4) directly, the following equation is obtained:

$$\zeta_i = -\Psi_e + \Psi_c \tag{5}$$

Assuming the conducting surface is initially uncharged, then the total charge on the surface should be zero:

$$\int_S \zeta_i dA = 0 \tag{6}$$

where S is the conducting surface, and A is the area of surface S.

Inserting Eqs. (5) and (6) leads to:

$$\Psi_c = \frac{\int_S \Psi_e dA}{A} \tag{7}$$

This is a numerical method to estimate the local induced zeta potential ζ_i on the conducting surface [19]. The zeta potential at the non-conducting channel wall ζ_w is constant and the value depends on the properties of the solid material and the liquid solution. Ψ_c is constant for an electrical conducting surface in an electric field. Since Ψ_e varies through the system, the induced zeta potential ζ_i along the conducting surface is not uniform. According to Helmholtz–Smoluchowski formula [17], the electroosmotic flow or the slip velocity is

$$\vec{U} = -\frac{\varepsilon_0 \varepsilon_r \zeta_i}{\mu}\vec{E} \tag{8}$$

where ε_0 and ε_r are the dielectric constant in vacuum and the dielectric constant of the solution, respectively. Because the induced zeta potential ζ_i and the local electric field strength $\vec{E}$ are not constant, and the induced zeta potentials are of opposite signs in the induced dipolar EDL, the slip velocity is in a different direction in different parts around the conducting surface, and vortices are formed in this way.

The boundary conditions of the electric field are

$$\vec{n} \cdot \vec{\nabla}\Psi_e = 0 \quad \text{at the channel walls} \tag{9}$$

$$\Psi_e = \Psi_1 \quad \text{at inlet 1} \tag{10a}$$

$$\Psi_e = \Psi_2 \quad \text{at inlet 2} \tag{10b}$$

$$\Psi_e = 0 \quad \text{at outlets} \tag{10c}$$

2.2. Velocity field

The velocity field is governed by the continuity equation and the momentum equation

$$\vec{\nabla} \cdot \vec{U} = 0 \tag{11}$$

$$\rho\left[\frac{\partial \vec{U}}{\partial t} + \vec{U} \cdot \vec{\nabla}\vec{U}\right] = -\vec{\nabla}P + \mu \vec{\nabla}^2 \vec{U} + \vec{E}\rho_e \tag{12}$$

where ρ and μ are the density and viscosity of the liquid, $\vec{U}$ is velocity, ρ_e is the local net charge density, $\vec{\nabla}P$ is the pressure gradient. Considering a steady flow, and no pressure driven force, the boundary conditions are

$$P = 0 \quad \text{at inlets and outlets.} \tag{13}$$

$$\vec{\nabla}P = 0 \quad \text{in the whole flow field.} \tag{14}$$

Because the net charge density is non-zero only in the double layer, the body force $\vec{E}\rho_e$ is zero except in the double layers. The flow is driven by the interaction force between the applied electrical field and the net charge in the double layers. Generally, the