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Numerical Simulation of a Novel Electroosmotic Micropump for Bio-MEMS Applications

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Abstract: High lamination in microchannel is one of the main challenges in Lab-On-a-Chip's components like micro total analyzer systems and any miniaturization of fluid channels intensify the viscose effects. In chip-scale, the electroosmotic flow is more efficient. Therefore, this study presents a MEMS-based low-voltage micropump for low-conductive biological samples and solutions, where twelve narrow miniaturized microchannels designed in one unit to efficiently using the electroosmotic effects which generated near the walls. Four microelectrodes are mounted in lateral sides of the microchannel and excited by low-voltage potential to generate pumping process inside the channel. We sweep the voltage amplitude and a linear variation of fluid velocity achieved by Finite-Element-Method (FEM) simulation. We obtain a net average velocity of 0.1 mm/s; by applying 2 V and -2 V to the electrodes. Therefore, the proposed low-voltage design is able to pumping the low-conductive biofluids for conventional lab-on-a-chip applications. *Copyright © 2014 IFSA Publishing, S. L.*

Keywords: Lab-On-a-Chip, MEMS-based micropump, Miniaturized narrow microchannel, Electroosmotic effect.

1. Introduction

Recently, advances in microfluidic devices have been employed in chemical and biological applications. Point of Care (POC) diagnosis and Lab-On-a-Chip (LOC) devices are being used to precise control and manipulation of small scale volumes of fluids in micro channel [1-2]. They also integrate a number of microfluidic components, such as pumps, valves, separators, mixers, reactors and detection system inside a single chip. These devices offer the ability to smaller fluid volumes, biocompatibility, low cost and shorter analysis time in the range of several to a few tens of microns. Micro-scale turns the viscous effects into the dominate factors [2-3]. Micropump is one of the important components of the microfluidic for on chip analysis.

Reducing the channel dimensions to the micron scale results in a laminar flow, characterized by low Reynolds number:

$$\mathrm{Re} = \frac{\rho U d_h}{\mu}, \qquad (1)$$

where *Re* is the Reynolds number, ρ is the fluid mass density, U is the average net flow velocity, μ is the dynamic viscosity of the fluid and d is the hydraulic diameter of the micro channel [4]. A micro pump can be applied in drug delivery systems [5], printer inkjet [6], electronic cooling systems [7] (VLSI systems, Laptops) and chromatography. Recently various electrokinetic micropumps have been proposed by researchers. Electrokinetic micropumps are popular because of their simplicity, low cost, simple fabrication, no need to the valves and integration beside the other microfluidic components such as micromixers, sensors and microseparators. As discussed in [8], the Electric Double Layer (EDL) is formed at the interface between solid walls and electrolytes. The solid surface becomes polarized and the counter ions coming from the bulk liquid shield this surface. In equilibrium state, the electrostatic attraction between the charged surface and the counter ions is balanced by thermal disturbing. The electrical double layer is divided into the Stern layer and Gouy-Chapman diffuse layer [8]. The Stern or inner layer is formed of ions absorbed onto the wall, while the ions of the Gouy-Chapman layer acts as diffuse layer. The separation plane between these two layers is called the shear plane and the potential at shear plane is zeta potential which it is function of pH [9]. The characteristic thickness of the Debye length λ_D is given by:

$$\lambda_D \approx \sqrt{\frac{\varepsilon D}{\sigma}}, \tag{2}$$

where the Debye length can be expressed in terms of the diffusion coefficient D; σ is the conductivity, and ε is the dielectric constant of the fluid [5]. A non-uniform ac electric field inside the microchannel acts as ACEO force. The normal component of the electric field charges the electric double layer at the solid-electrolyte interface, while the tangential component of electric field produces a force on the induced charge in the diffuse layer and the fluid is set into motion, which will be discuss schematically in theory section. Nowadays, two-phase, three-phase and also four-phase electrode arrays have been employed in ACEO micropumps.

Electrode arrays with 2-phase have simple fabrication process but small pumping velocity which occurs only at the specific frequency. In three-phase (3-phase) and four-phase (4-phase) electrode arrays flow direction can change easily by swapping phase between their electrodes. However, the electrode arrays with 3-phase and 4-phase require complicated fabrication due to bridging structures. Ajdari [10] demonstrated theoretically feasibility of creating an ACEO-induced pumping effect that uses asymmetric electrode arrays. Two major techniques have been used for driving microfluidic flows: using coplanar electrode arrays, such as asymmetric electrode pairs subjected to common ac signal [11], or electrode arrays with equal width subjected to a three or four phases traveling-wave potential [12]. García-Sánchez [13] demonstrated that the travelling-wave method is more efficient because of higher velocity can be achieved with low voltage. Experimental evidence has shown that increasing the voltage amplitude can cause higher pumping velocity in the planar electrode structures. However, by increasing voltage beyond a certain critical value, undesirable effects, such as electrolysis, electrode degradation and hydrodynamic instabilities have been observed. Urbanski experimentally investigated the effects of the step height on the pumping velocity induced by asymmetric pairs of 3D electrodes and compared them with the obtained results from a simulation model [14]. García-Sánchez [15] theoretically studied the effect of electrode height on the performance of TWEO micropumps. As a result, the net pumping velocity in a 3D electrode array can be increased as much as 2.7 times a 2D structure. Vafaie and *et al.*, presents LOC components including micromixer, micropump and microseprator by using electroosmtic force [16-18].

The electroosmotic effect arises from the solid/electrolyte interface. Therefore, in this research we presents a novel low voltage micropump by using multiple narrow microchannel and investigate the pumping operation in the miniaturized microchannel of conventional Lab-On-a-Chip applications. For this purpose, firstly the geometrical design of the micropump will describe in Section 2. The theory and boundary conditions of the fully-coupled electroosmotic micropump are discussed in Section 3. Section 4 will perform a set of simulations to extensively investigate the induced electric field, velocity profile and voltage effect on pumping rate. Conclusion and comparison with other micropumps will be discussed in Section 5.

2. Design Consideration

In order to overcome the high driving voltages of conventional electroosmotic micropumps, we propose a new design. As it can be seen from Fig. 1, the geometrical model contains units of single microchannel. Where, a single wide microchannel consists of twelve narrow channels and the actuation electrodes mounted on the lateral walls of wide microchannel. The values of geometrical parameters of the proposed micropump were assigned in accordance with Table 1.

Table 1. Geometrical parameters of microchannel for pumping process.

Symbol	Description	Value [µm]
W_{inlet}	Channel Inlet Width	300
W_{outlet}	Channel Outlet Width	250
$L_{channel}$	Channel Length	700
L_{elec}	Electrode Width	300
W_1	Shown in Fig. 1	265
W_2	Shown in Fig. 1	150
W_3	Shown in Fig. 1	200
W_4	Wide channel width	285
W_5	Shown in Fig. 1	315
W_6	Shown in Fig. 1	250
r	Radios of curvature	150

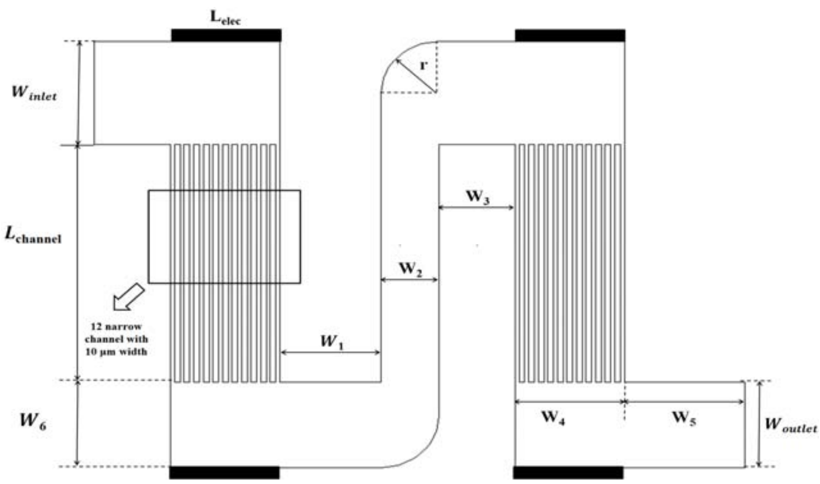

Fig. 1. Two-dimensional Schematic view of microchannel and narrow microchannels, four microelectrodes are placed in the top and down of microchannel for pumping process.

The system will be able to generate an electrosmotic force near the twelve narrow channels by actuating the electrodes with proper low-voltage electric field. Arraying multiple unites of these microchannels, will allow us to design a micropump with an effective pressure source. Since the length, width, and depth of geometrical model are so much larger than the microelectrode's height, the electrodes can be modeled as flat line in simulation modeling [19].

3. Theory and Boundary Conditions

The concept of electric double layer formation is because of electrochemical effects. Electrochemical equilibrium between a solid surface and an electrolyte solution typically leads to the interface acquiring a net fixed electrical charge, a layer of mobile ions, known as an EDL, forms in the region near the interface [20-21]. These effects are schematically illustrated in Fig. 2.

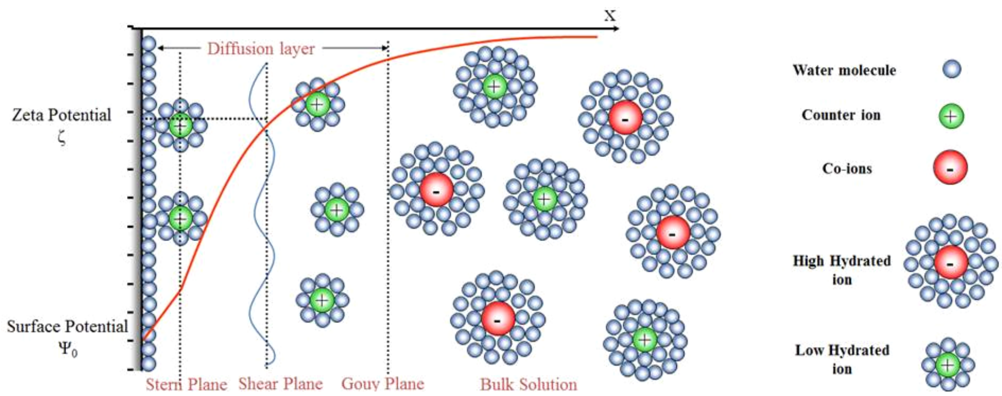

Fig. 2. Electric double layer formation; the whole ions in bulk solution are highly surrounded by water molecules, so the bulk solution is electrically neutral. However, due to the electrochemical effect at surface, the ions are low hydrated (by water molecules) near the solid-wall/electrolyte interface. So, a thin layer charged by counter-ions, at interface between the solid wall and electrolyte.

By applying a tangential electric field to the electrolyte solution, the charges in the electrical double layer at the interface of electrolyte solution and surface of electrodes, experience a considerable force. Therefore, these EDL charges move and as a result pull the bulk fluid along the actuation path, which cause to pumping effect. For analyzing the pumping operation we solve a mulyiphysic problem including:

a) The fluidic Incompressible Navier-Stokes equation and continuity equation,

b) The electric field inside the microchannel by electrostatic equation.

3.1. The Fluid Behavior

Both Navier-Stokes equation (Equation (3)) and continuity equation (Equation (4)) [19], govern the incompressible liquid flow:

$$\rho\left[\frac{\partial \bar{u}}{\partial t}+\vec{u}\nabla\bar{u}\right]=-\nabla p+\mu\nabla^{2}\vec{u}+f_{E}, \qquad (3)$$

$$\nabla \overline{u} = 0, \quad (4)$$

where ρ is the fluid density, u is the net flow velocity, p is the pressure in the micro channel, μ is the fluid viscosity, f_E is the driving electric force, which generated by external electric field in micro channel. Electrical driving force represents interaction between electrical double layer (EDL) and excess ions [8].

3.2. The Electrostatic Model

From electrical point of view, we actuate the electrodes by electric potential of *f(t)*:

$$\begin{aligned} f(t) &= V_0 \quad \text{at Upper Electrodes} \\ f(t) &= -V_0 \quad \text{at Lower Electrodes} \end{aligned} \quad (5)$$

As a result of applied electric field, the ions in the EDL experience a tangential force. In such conditions, the highly miniaturized narrow microchannels improve the electroosmotic velocity near the channel walls. The electroosmotic velocity, U_{eo}, is well approximated by Equation (6) and known as the Helmholts-Smoluchowski equation which is valid for thin double layers [23].

$$U_{eo} = -\frac{\varepsilon \zeta E}{\mu}, \quad (6)$$

where $\varepsilon = \varepsilon_0 \varepsilon_r$, ε_0 is the dielectric permittivity in a vacuum, ε_r is the relative dielectric permittivity of the liquid, ζ is the electrokinetic zeta potential, and E is the electric field.

4. Results and Discussion

The main aim of this study is to investigate the pumping effects with low voltage and corresponding low power in portable biomedical and chemical LOC and POC devices. Fluid velocities are computed using finite element method. To simulate the time-averaged velocity inside the microchannel the water was used as the working fluid. The required properties of water and other physical material properties are listed in Table 2 [10, 21]. As discussed before, due to microscales and corresponding very low Reynolds number inside microchannel, the system is highly laminar.

Table 2. Geometrical parameters of a mixer unit.

Symbol	Description	Value
σ	Electric conductivity	0.001 [S/m]
ε_r	Dielectric Constant of Water	80.2
ρ	Mass Density of Water	1000 [kg/m^3]
μ	Dynamic Viscosity of Water	0.001 [Pa.s]
ξ	Zeta Potential	- 80 [mV]
V_0	Amplitude of Electric Potential	From 0 to 10 [V]

In simulation model, the pressures at the inlets and outlet of the microchannel were specified as zero and the pressure gradient at the channel walls is set to be zero, which considering no flux across the walls:

$$\begin{aligned} P &= 0. \\ n\nabla P &= 0 \end{aligned} \quad (7)$$

It assumes that the pumping operation do not affect the working fluid properties such as fluid dynamic viscosity and μ, mass density, ρ [1, 11]. The four electrodes are excited by electric potential of *f(t)* as discussed in equation (5) to generate electroosmotic effect inside the microchannel. The most important boundary conditions of pumping process are schematically indicated in Fig. 3.

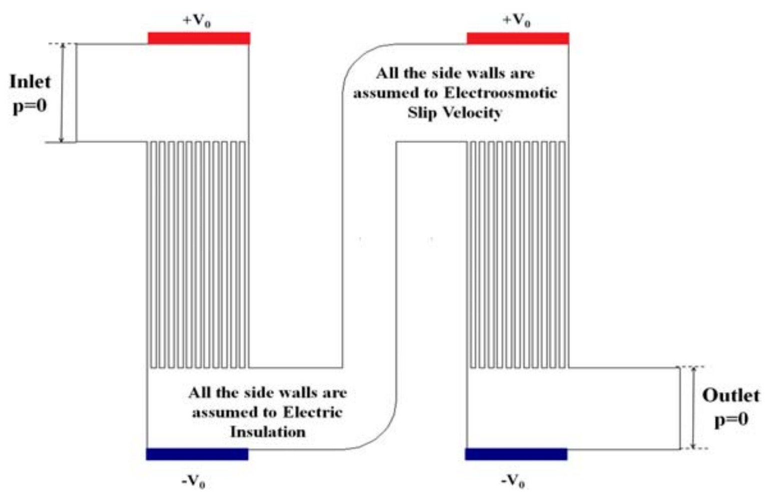

Fig. 3. Micropump boundary conditions for both Fluid flow and electric field.

A set of simulations were done to extensively investigate the pumping process. As illustrated in Fig. 4 the electric potential of 10 Volt and -10 V are applied respectively to the top and down electrodes. As a result, an electric field is induced inside the channel ($\overline{E} = -\nabla V$). As a prove of pumping effects inside the microchannel, Fig. 5 illustrates the generated velocity field arrows, streamline and surface plot, where the maximum pumping velocity of 1.1 mm/s achieved by applying 10 V to the V_0. Actually, the miniaturized narrow microchannels enhance the pumping effect by efficiently using the electroosmotic effects near the multiple narrow channels. It should be noted that the pumping rate of 0.1 mm/s is conventional for miniaturized lab-on-a-chip applications [22, 24], and the proposed micropump are able to achieve average net velocity of 0.1 mm/s with using very low amplitude voltages.

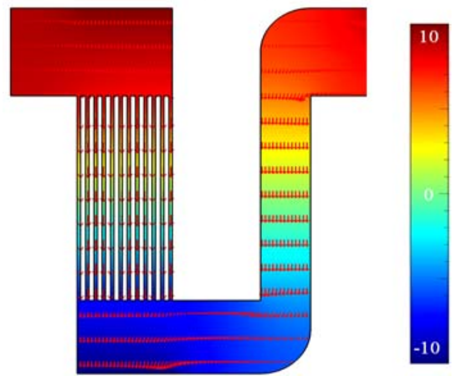

Fig. 4. Electric potential surface plot and correspomding generated uniform electric field inside the channel, indicated with red arrows.

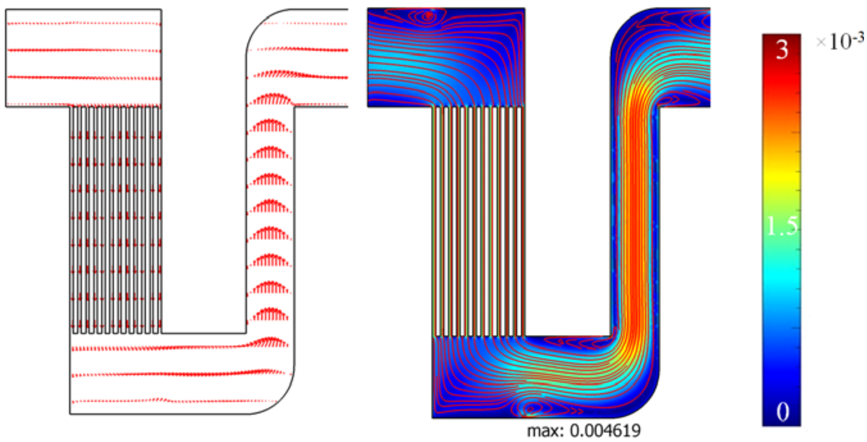

Fig. 5. Velocity field inside the microchannel; including arrows, streamlines and surface plot.

As mentioned in previous section, the external electric field acted on the electric charges in the electric double layer. The tangential component of the electric field created a driving force that pulling the fluid inside the channel, the pumping velocity (netflow) is defined as [25]:

$$U = \frac{1}{L} \int_{x=0}^{x=L} u dx \quad (8)$$

4.1. Voltage Effect

The velocity profile are indicated in Fig. 6A, which has a parabolic profile with maximum velocity of 1.1 [mm/s]. it should be notted that the electroosmotic pumping effect is largely affected by electric conductivity of solution. Actually, the tangential component of electroosmotic force reduces by decreasing the EDL thickness, λ_D. Also, the electric double layer thickness compressed by increasing the electrical conductivity of fluid [26, 27]. We sweep the electric potential V_0 from 0 to 10 and as indicated in Fig. 6B a linear increasing in net velocity observed. This effects allow us to control the pumping velocity with variation of electric potential.

5. Conclusions

This study investigates numerically a novel electroosmotic micropump, where multiple narrow microchannels are used for enhancing the electroosmotic effect. The electroosmotic force arise from the channel walls. Therfore we improve the electroosmotic effect by increasing the channel walls. The pumping performance is studied by applying low voltages on the electrodes. The result of this study reveals that the pump operates with low voltage which it is of interest in portable applications. This device is effective for fluid buffer with low electric conductivities, where we have thick electric double layerr. We study the fully-coupled multi-physic micropump by Finite element method (FEM) and linear relation achieved between the applied electric potential and pumping rate. For example, a pumping rate of 0.2 mm/s is achieved for 4 [V] voltage which is very good for low power portable applications and avoid electrolysis. This micropump is compared with other micropumps in accordance with Table. 3.

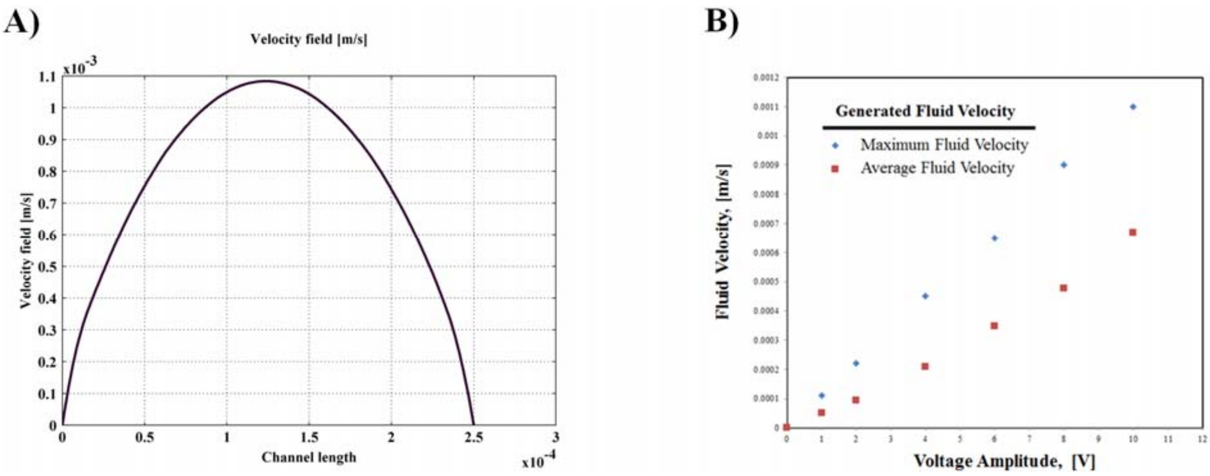

Fig. 6. Velocity results; (A) velocity profile at outlet and (B) voltage variation effect on pumping velocity.

Table 3. Comparison table.

References	Type of Pump	Channel width [μm]	Fluid Electric conductivity [S/m]	Applied Voltage [V]	Typical velocity
[15]	3D ACEO	100	KCl	3	1.2 [mm/s]
[15]	TWEOP	-	1e-3, KCl	2	38 [μm/s]
[28]	ACEO	1000	0.0015, KCl	4.5	37 [μm/s]
[29]	3-phase TWEOP	-	1.3e-7, Ethyl ethanol	8	270 [μm/s]
[30]	3D ACEO	500	-	2	190 [μm/s]
This work	Electroosmotic with multiple narrow channels	250	1e-3, Water	4	210 [μm/s]

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