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Electrode Engineering for On-Chip Liquid Driving by Using Electrokinetic Effect

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Abstract**—**High lamination in microchannel is one of the main challenges in on-chip components like micro total analyzer systems and lab-on-a-chips. Electro-osmotic force is highly effective in chip-scale. This research proposes a microfluidic-based micropump for low ionic strength solutions. Narrow microchannels are designed to generate an efficient electroosmotic flow near the walls. Microelectrodes are embedded in the lateral sides and actuated by low electric potential to generate pumping effect inside the channel. Based on the simulation study, the fluid velocity increases by increasing the electric potential amplitude. We achieve a net flow velocity of 100 µm/s, by applying +/- 2 V to the electrode structures. Our proposed low voltage design is of interest in conventional lab-on-a-chip applications.

Keywords**—**Integration, electrokinetic, on-chip, fluid pumping, microfluidic.

I. Introduction

THE progress in integrated electronic circuits encouraged the researchers to employ the integration concept into variety of fields. 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 microfluidic components, like valves, valves, mixers, separators, pumps and reaction system inside a chip. These components offer the ability to less sample solution, compatibility, and shorter testing time. Micron scale turns the viscous effect into the dominate factor [2], [3]. Micro-liquid driving is one of the important devices of the microfluidic for biological analysis. Reducing the dimensions to the micro /nanoscale results in a creeping flow, characterized by low Reynolds number (*Re*):

$$\mathrm{Re} = \frac{Inertial \quad Effect}{Vis\cos e \quad Effect} = \frac{u.d_h}{v}. \tag{1}$$

where ρ is the fluid mass density, µ is the dynamic viscosity of the fluid, U is the average velocity and d is the diameter of the channel [4]. Micropump can be employed in drug delivery devices [5], inkjet printers [6], cooling systems such as laptops [7] and liquid chromatography. Electrokinetic micropumps have been widely proposed by literatures and researchers. Electrokinetic micropumps are conventional because of simple fabrication, simplicity, low cost; it does not need to the valves and integration beside the other microfluidic devices likes sensors and microseparators. As presented in [8], Electric Double Layer (EDL) is shaped at the liquid-solid interface. The solid walls become polarized and the counter ions coming from the bulk solution start to shield this surface. The electrostatic attraction between the charged surface and ions is balanced by thermal effects. The EDL 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 act as diffuse layer. The separation plane between these two layers is called the shear plane. The potential at shear plane is called zeta potential which is function of pH [9]. The characteristic thickness of the EDL λ_D is formulized by:

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

where D is the diffusion coefficient, σ is the conductivity, and ε is the dielectric constant of liquid [5]. A time-varying ac electric field inside the microchannel acts as AC electroosmotic force. The normal component of AC electric field charges the EDL at the solid-liquid interface, while the tangential component of AC electric field causes to liquid motion on the induced charge in the diffuse layer. Two-phase, three-phase, and also four-phase electrode arrays have been employed in AC electroosmotic pumps. Electrode structures with two-phase have simple manufacturing process but small pumping velocity which leads only at the low frequencies. In the case of three-phase and four-phase electrode structures, flow direction can be changed easily by switching AC potential phase between the electrodes. However, the electrode arrays with three-phase and four-phase require complicated fabrication. Ajdari [10] presented a feasibility of creating an electroosmotic-induced pumping effect that uses asymmetric electrode structures. Two main methods have been applied for driving liquids in microchannel: using electrode arrays, such as asymmetric electrode pairs subjected to AC signal [11], or electrode arrays with equal width subjected to three and four-phase traveling-wave signal [12]. Reference [13] showed that travelling-wave mechanism is more effective, and high amplitude velocity can be achieved by applying a low

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voltage17fr060493. Experimental results have discovered that increasing the electric potential amplitude can cause higher fluid velocity by using the planar electrode structures. By increasing the voltage, bad effects such as electrolysis and electrode degradation have been observed. Urbanski et al. investigated the effects of the step height on the pumping velocity induced by asymmetric pairs of three dimensional electrodes [14]. Reference [15] studied the effect of electrode height theoretically on the performance of traveling wave electroosmotic pumps. The net liquid velocity in a three-dimensional electrode structure can be increased as much as 2.5 times a two-dimensional structure. Vafaie et al. present LOC devices including mixer, pump, and manipulator by using electro-osmotic force [16]-[18].

ACEO force arises from the solid/liquid interface. Therefore, in this study, we investigate a novel low voltage liquid driving by using multiple narrow microchannel and proposing the pumping operation in the more miniaturized and integrated microchannel for biological applications.

II. Material and Methods

A. Design

In order to decrease the high actuation voltages of conventional electroosmotic pumps, we propose a new form of microchannel and electrode design. As it is shown in Fig. 1, the geometrical model contains units of microchannel. A single wide microchannel including of multiple narrow channels and the actuation electrodes is embedded on the lateral walls of microchannel. Dimensional values of geometrical parameters were assigned in accordance with Table I. The proposed design is able to generate an AC electro-osmotic force near the narrow channels by properly actuating the electrodes by low-voltage. Arraying multiple units of the channels will allow us to propose a liquid driving system with an efficient pressure source. The length, width, and depth of proposed model are much larger than the microelectrode's height; therefore, the electrodes can be modeled as embedded electrode structure in simulation analysis [19].

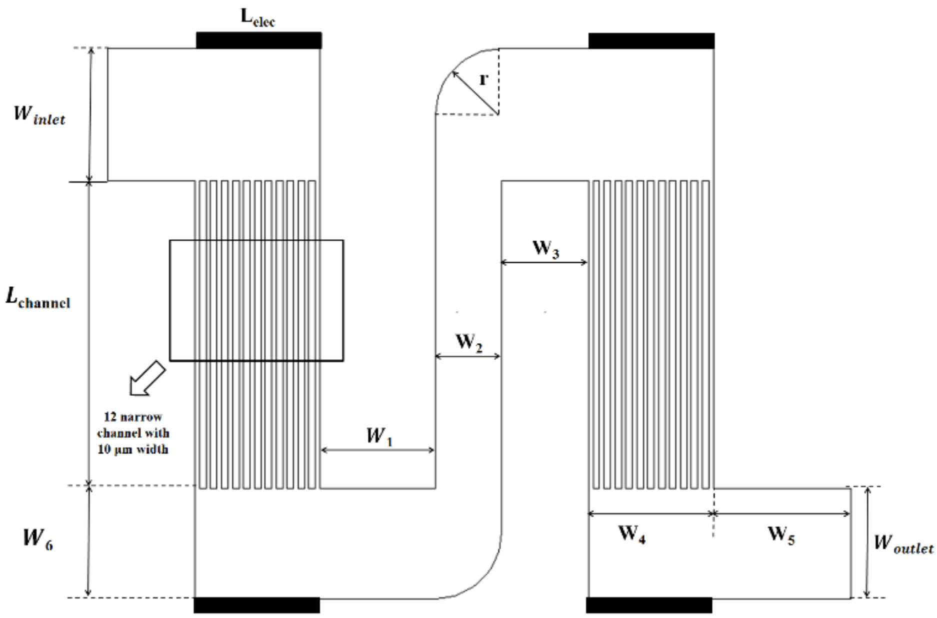

Fig. 1 Two-dimensional view of entire channel and the narrow channels, four pieces of electrodes are placed in the top and down of channel for liquid driving process

TABLE I
GEOMETRICAL PARAMETERS OF MICRO-CHANNEL FOR PUMPING PROCESS

Symbol	Description	Value
W_{inlet}	*Channel inlet width*	*300*
W_{Outlet}	*Channel outlet width*	*250*
$L_{Channel}$	*Channel length*	*700*
$L_{Electrode}$	*Electrode width*	*300*
W_1	*Shown in Fig. 1*	*265*
W2	*Shown in Fig. 1*	*150*
W3		*200*
W4	*Wide channel width*	*285*
W5	*Shown in Fig. 1*	*315*
W6	*Shown in Fig. 1*	*250*
r	*Radios of curvature*	*150*

B. Theory

The concept of EDL formation is due to electrochemical effects. Electrochemical equilibrium between a solid surface and an liquid solution 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]. Both Navier-Stokes equation and continuity equation [19] govern the incompressible liquid flow:

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

TABLE II
MATERIAL PROPERTIES FOR WORKING FLUID AND PHYSICAL EQUATIONS

Symbol	Description	Value
ρ_m	density of fluid	1×10^{-3} [N.s/m^2]
μ	viscosity of fluid	1×10^{3} [kg/m^3]
σ	electric conductivity of fluid	1 [mS/m]
ζ	Zeta potential	-80 mV
ε_r	dielectric constant of Fluid	80.2
V_0	Applied electric potential	From 0 to 10 V

$$\nabla \bar{u} = 0. \tag{4}$$

where u is the net flow velocity, ρ is the fluid density, f_E is the driving electric force, p is the pressure in the micro channel, μ is the fluid viscosity. Liquid driving force indicates interaction between EDL and excess ions [8]. From electrostatic view, we drive the electrodes by applying an electric potential of *f*(t):

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

As a result of electric field actuation mechanism, the ions at the EDL experience a tangential force. Therefore, the miniaturized narrow channels improve the liquid velocity near the channel walls. The electroosmotic velocity, U_{eo}, is approximated by (6) and called as the Helmholts-Smoluchowski equation (valid for thin double layers) [23].

$$U_{eo} = -\frac{\varepsilon \zeta E}{\mu}. \tag{6}$$

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

III. RESULTS AND DISCUSSIONS

The basic goal of this research is to investigate the liquid driving effects by low voltage actuation and corresponding low power in portable biomedical and chemical LOC devices and point-of-care systems. Liquid driving velocities are analyzed by using COMSOL Multiphysics (based on finite element method). To simulate the velocity inside the channel, the water solution was used as liquid working fluid. The properties of liquid and physical material properties of system are listed in Table II [10], [21]. As discussed earlier, because of microscale limitations and small Reynolds number inside the channel, the system is completely laminar. In numerical model, pressures at the channel inlet and outlet are specified as zero, and the pressure gradient at the channel walls is fixed to zero (considering no flux across the solid surfaces):

$$\begin{aligned} P &= 0. \\ n\nabla P &= 0 \end{aligned} \tag{7}$$

It assumes that the liquid driving operation cannot affect the working fluid properties [1], [11]. The electrode structures are actuated by electric potential *f(t)* to induce AC electroosmotic force inside the channel. Most important boundary conditions of liquid driving process are visualised in Fig. 2.

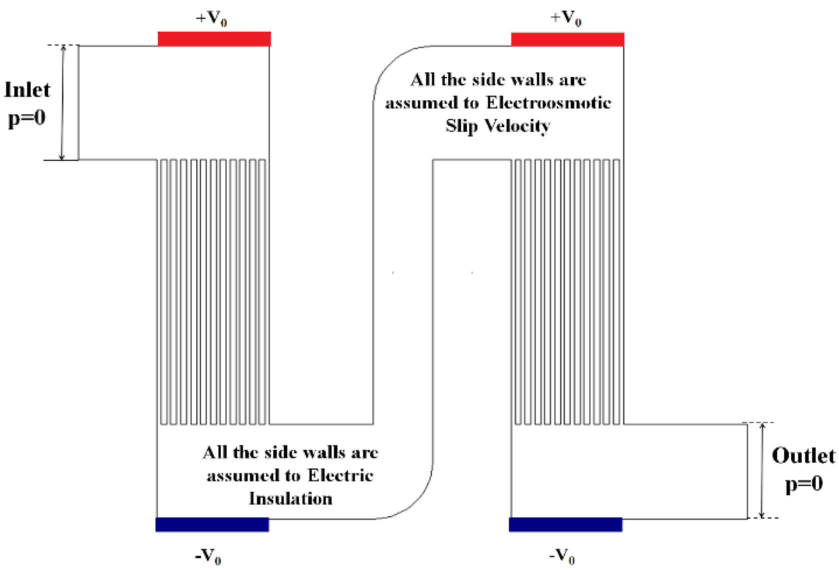

Fig. 2 Micropump boundary conditions for both Fluid flow and electric field

A set of numerical studies were done to broadly investigate the liquid driving process. As visualized in Fig. 3, the electric voltages of 10 V and -10 V are applied to the top and down electrodes, respectively. As a result, an electric field is induced inside the channel ($\bar{E} = -\nabla V$). As a prove of fluid pumping effects, Fig. 4 indicates the generated liquid velocity arrows, streamline and surface plot, where the maximum liquid driving velocity of above 1 mm/s is achieved by applying 10 V to the electrodes. The integrated narrow channels enhance the liquid driving effect by effectively using the AC electroosmotic actuation mechanism near the narrow microchannels. It should be noted that a liquid driving rate of 0.1 mm/s is completely conventional for LOC devices and point-of-care applications [22], [24], [25], and the proposed design is able to achieve net velocity of 0.1 mm/s by using low voltages.

The applied electric potential acted on the electric charges and ions in the EDL. It should be noted that the AC electroosmotic driving mechanism is largely affected by ionic strength of solution. Actually, the tangential component of electroosmotic force reduces by decreasing the EDL thickness, λ_D. Also, the EDL thickness compressed and collapsed by increasing the electrical conductivity of fluid [26], [27]. We swept the electric voltage from 0 to 10 V, and as shown in Fig. 5 a linear increasing profile is observed in net velocity amplitude. These effects allow us to control the liquid pumping velocity with variation of electric voltage.

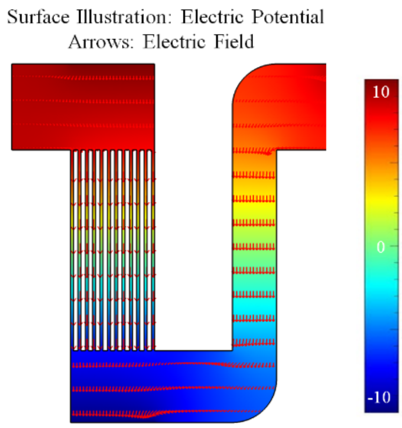

Fig. 3 Electric voltage surface plot and the corresponding generated uniform electric field inside the channel, indicated with red arrows

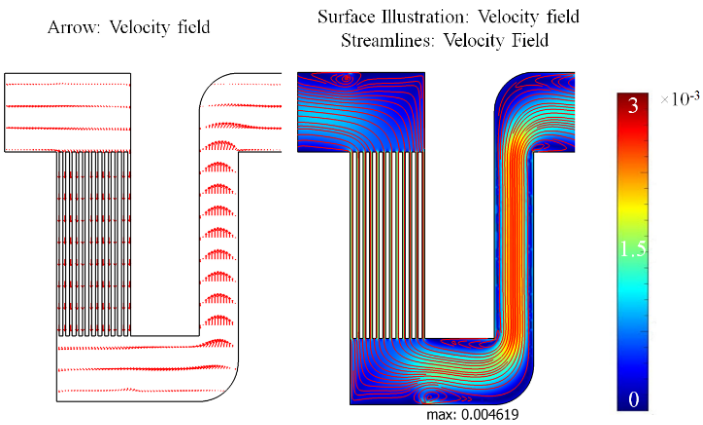

Fig. 4 Velocity field inside the channel; including arrows, streamlines and surface plot

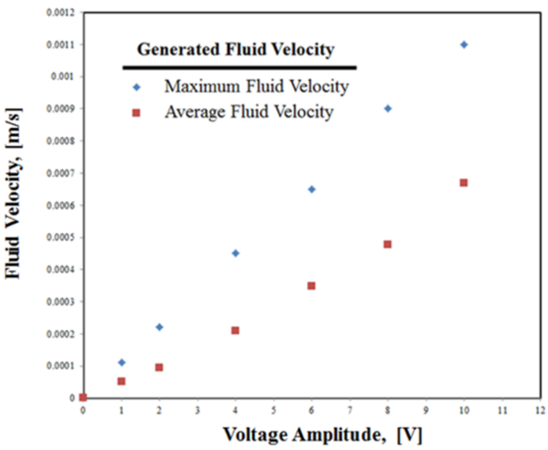

Fig. 5 Velocity results; voltage variation effect

IV. Conclusion

This research investigates simulation study of a novel AC electroosmotic liquid driving effect, where multiple narrow microchannels are used for enhancing the electrokinetic effect. AC electroosmotic force starts to arise from the channel walls. Therefore, we improve the AC electroosmotic effect and liquid driving force by increasing the channel walls. The driving efficiency is studied by employing low electric potentials over the electrode structures. The result showed that the proposed system operates with low power consumption and electric potential; and it is of interest in portable applications. This microdevice is efficient for liquid solution and buffer with low electric conductivities, where the system is characterized by thick EDL. The Finite element method (FEM) multi-physic coupled liquid driving system is studied by COMSOL and linear relation achieved between the applied electric voltage and liquid pumping rate.

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