Design and Fabrication of a Flow Delivery Microdevice with Asymmetric Microelectrodes Pairs

Hong Kiat Tay1,2, Daniel Lee*1, Guolin Xu1 and Chun Yang2
1Institute of Bioengineering and Nanotechnologies, 31 Biopolis Way, The Nanos, #04-01, Singapore 138669
2Department of Mechanical and Aerospace Engineering, Nanyang Technological University, 50 Nanyang Avenue, Singapore 639798
E-mail: yslee@ibn.a-star.edu.sg

Abstract. A microdevice using AC electrokinetic flow and negative dielectrophoresis for cell delivery is proposed, designed and fabricated. The device is made from an unequal width interdigitated microelectrode located at the bottom of microchannels, with AC voltage applied to the small and large electrode. Negative dielectrophoresis is used to prevent the cells from sinking down to the bottom of delivery channel. The flow rate is affected by the electrode pair dimensions, applied frequency and voltage and are analysed. The device electrode width is from 4μm to 20μm and it is microfabricated on glass wafer using gold thin film. 3 microchannels connected to 2 reservoirs are covered by the electrode pair for the delivery channel. SU-8 micromold is used for the channels’ PDMS fabrication.

Keywords: Electroosmosis, Microfluidics, Dielectrophoresis, Cell flow delivery.

1. Introduction
“Lab-On-a-Chip”, a miniature bio/chemical or medical diagnosis system using a small footprint and volume, is improving in its efficiency and throughput. In such a system, fluid delivery, mixing, sample separation/concentration and detection are achieved inside the device[1]. One of the challenges to achieving automated total analysis is to deliver small amounts of biological samples and reagents through the channels. Micropumps able to create pressure differences between the ends of the channel have been proposed and integrated in such a system. [2] However, such micropumps are too complex and expensive for the disposable biological application. A technique that is simple, cheap and reliable for sample and reagent delivery is highly desirable.

Electroosmotic flow is fluid motion produced by electric field acting on the net fluid charge, which is produced at the fluid-solid interface. Advantages for using electroosmotic flow for pumping solution includes: no moving parts hence it is reliable, simple in structure, high flow rate with low power (1 VRMS/5-10KHz for 450μm/s flow) [3]. It has been reportedly used in chromatography. [4]

In this paper, we present a device using alternating current (AC) electroosmotic flow for cell delivery. Applying electric fields generated by an asymmetric electrode array, the device combines
dielectrophoresis (DEP) for floating the cells and electroosmotic flow for moving the cells along a channel. The principle and factors that affect the cell flow are presented and discussed. The microfabrication of the device is detailed.

2. Working principle
AC electroosmosis is an electrokinetic phenomenon at frequencies below 1MHz. As shown in Figure 1, an AC electroosmosis device consists of an interdigitated array of unequal width electrodes located on the bottom of a microchannel.

AC voltage is applied to the two electrodes. The electric potential within the electrode will cause charge to accumulate on the electrode surface, which will change the charge density near the surface and forms an electric double layer as shown in the Figure 1. The electric field generated among the electrodes causes the ions in the double layer to move towards one electrode or the other. A net coulombic force is generated on the double layer above the electrodes due to the tangential component of the electric field. This force will cause fluid movement.

Figure 1: Principle of AC electroosmosis

Figure 2 shows the forces applied on a cell during delivery. Cells may sink down to the bottom of the device channel due to the gravity $F_{SD}$, or they will be attracted to the electrode edge due to dielectrophoresis (DEP), $F_{DEP}$ if the applied frequency results in positive DEP. The cells also experience a dragging force, $F_g$ from the viscosity of the moving medium it is submerged in.

The time averaged dielectrophoresis force applied to cell can be expressed as [5]:

$$F_{DEP} = 2\pi \varepsilon_p \varepsilon_m r^3 \text{Re}[f_{CM}] |E_{RMS}|^2$$

with $f_{CM} = \frac{\varepsilon_p - \varepsilon_m}{\varepsilon_p + 2\varepsilon_m}$

$\varepsilon_p^*$ and $\varepsilon_m^*$ are the complex permittivity of the cell and the fluid respectively.

Figure 3 shows the electric field distributions along the channel by using MAXWELL™ electric field simulation software. The electrodes simulated in this case were 20μm and 5μm for the large and small electrodes respectively, using 1Vpp. The closer to the electrode surface, the higher the electric...
field strength. Most mammalian cells will exhibit negative DEP properties in the low frequency region. This means the DEP force will push the cell to move up and against the gravitational force; hence cells will not stick to the channel bottom.

3. Device design

Numerical model for using asymmetric electrodes array for liquid delivery has been developed by Ajdari [5]. Brown et al. [6] has developed a simplified 2D model to describe electroosmotic flow. Based on this model, the large electrodes have width $L$ and the small electrode width $S$, the gap between them is $G$, as illustrated in Figure 1. In order to simplify the algebra, a variable $x$ describes the position of both ends of a conduction path across the surface of the electrodes. Let the ratio of the electrode widths (i.e. $L/S$) be $k$. Let an origin be chosen between the electrodes such that a conduction path that starts at a distance of $x\sqrt{k}$ from this origin over the large electrode will end at a point $x\sqrt{k}$ from this origin over the small electrode. Let $x_{\text{min}}$ and $x_{\text{max}}$ be chosen such that the large electrode lies between points $x_{\text{min}}\sqrt{k}$ and $x_{\text{max}}\sqrt{k}$ from the origin and the small electrode lies between points $x_{\text{min}}/\sqrt{k}$ and $x_{\text{max}}/\sqrt{k}$ from that same origin. This means that $x_{\text{min}}$ is equal to $G/(\sqrt{k} + 1/\sqrt{k})$ and $x_{\text{max}}$ is equal to $(G + L + S)/(\sqrt{k} + 1/\sqrt{k})$. The average velocity is given as (modified from [4]):

$$V_{\text{ave}} = \frac{\int_{x_{\text{min}}}^{x_{\text{max}}} v_D(x)dx}{x_{\text{max}} - x_{\text{min}}} = \frac{\psi_o V_o}{2(x_{\text{max}} - x_{\text{min}})} \left\{ \frac{\left(\omega \sqrt{x_{\text{min}}x_{\text{max}}} / \epsilon_o \right) \left( x_{\text{max}} - x_{\text{min}} \right)}{\left(\omega \sqrt{x_{\text{min}}x_{\text{max}}} / \epsilon_o \right) + \frac{x_{\text{min}}}{x_{\text{max}}} \left(\omega \sqrt{x_{\text{min}}x_{\text{max}}} / \epsilon_o \right) + \frac{x_{\text{max}}}{x_{\text{min}}} \right\}$$

Where $\omega_o = \frac{2\lambda_D \sigma}{\epsilon \pi x}$, $\lambda_D$ is the Debye layer, $\sigma$ is the conductivity of the solution, $\epsilon$ is the permittivity of the solution, $\psi_o$ is the applied voltage.

Using this model, we have predicted the effects of the large and small electrode dimensions, the applied frequency and the applied voltage as shown in Figure 4 (a)-(c) respectively.
4. Microfabrication

The 3D model of the device is shown in Figure 5. It contains a microelectrode pair on the glass surface. A micro moulded PDMS cover forms 2 reservoirs and 3 channels. Table 1 shows the large and small electrode sizes and the channel sizes used in this work.

The electrode array was made by depositing 50nm of chromium as seed layer on a Pyrex 7740 glass wafer followed by depositing 200nm of gold by using an electron beam evaporation machine from CHA. Gold was chosen, as it does not form oxides layer in water, which would add to the surface capacitance when it is being used. The patterning of the electrodes is done by standard micromachining process using a precision chromium glass mask. Briefly, evaporate-coat HDMS as an adhesion promoter on the gold layer, followed by spin-coating AZ7220 photo resist on its surface. After soft-baking at 110°C for 1 min on a hot-plate, the wafer is exposed to UV light in mask aligner (EVG620) for 10 seconds to expose the mask pattern. After post-baking on a hotplate at 100°C for 1 min, the wafer is then developed using AZ400k developer. The metal layer is etched by gold and chromium etchant respectively. Figure 6 (a) is the fabricated wafer with 8 different designs. Figure 6(b) shows the detail of the fabricated asymmetric electrode pairs. Dicing saw is used to singulate the fabricated wafer.
SU8-2050 photoresist on silicon wafer as substrate was used for molding the microchannels and reservoirs. PDMS is selected for its transparency, easy micromolding and ability to bond on glass surfaces. Before PDMS moulding, an anti-stick layer (tridecafluoro-1,1,2,2-tetrahydrooctyl trichlorosilane, from Aldrich) was applied onto the mould by vacuum evaporation method. The mold is then put inside a container. Two components of Sylgard184 Silicon Elastomer (from Dow Corning) was mixed and poured slowly into the silanized mold. The mold was then placed inside a vacuum dessicatior for about one hour to release air bubbles trapped inside the uncured PDMS mixture. The whole set up was then cured inside an oven at 70°C for an hour. The PDMS is cut and assembled with the microelectrode pair. Figure 6 (c) is the picture of the assembled device.

5. Conclusion
Micorfabricated cell delivery device with asymmetric electrode pair is proposed. The flow rates relationship to the electrode width, electrode gap, applied frequency and applied voltage are studied. Using negative dielectrophoresis, a method for preventing cell sinking or sedimentation on the delivery channel is proposed and studied. The device has been fabricated by using microfabrication technology.

References
[1] Ying, Huang, E. L. Mather, J. L. Bell and M. Madou. 2002. MEMS-based sample preparation for molecular diagnostics. Anal Bioanal Chem 372:49-65.
[2] R. H. Liu, J. Yang, R. Lenigk, J. Bonanno and P. Grodzinski, “Self-Contained, Fully Integrated Biochip for Sample Preparation, Polymerase Chain Reaction Amplification, and DNA Microarray Detection” Anal. Chem. 2004, 76, 1824-1831
[3] P. K. Wong, T-H. Wang, J. H. Deval and C-M Ho, “Electrokinetics in micro devices for biotechnology applications” IEEE transactions on mechatronics Vol 9, No.2 PP 366-376, 2004
[4] S. Debesset, C. J. Hayden, C. Dalton, J. C. T. Eijkel and A. Manz, “An AC electroosmotic micropump for circular chromatographic applications” Lab Chip, 2004, 4, 396-400.
[5] Jones T. B. Electromechanics of particles. Cambridge University Press. 1995.
[6] Armand Ajdari, “Pumping liquids using asymmetric electrode arrays” Physical review E, Vol 61, No.1 45-48, 2000.
[7] A. B. D. Brown, C. G. Smith and A. R. Rennie, “Pumping of water with ac electric field applied to asymmetric pairs of microelectrodes”, Physical review E. Vol 63, 016305, 2000.