

# A DIELECTROPHORESIS MICROPUMP FOR ON-CHIP PARTICLES TRAPPING AND BLOOD DRIVING IN A VIRTUAL CHANNEL

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## ABSTRACT

Continuous liquid pumping at a steady and tunable flow rate is demonstrated by creating a proper pressure difference across an electric-field-formed virtual microchannel using dielectrophoresis (DEP). Because the liquid is regulated with no physical channel walls or mechanical movable parts, the fabrication and packaging of the device is simple while the issues of liquid leakage and dead volume are eliminated. With appropriate electrode designs, particles suspended in the pumped liquid can be manipulated by DEP simultaneously. The continuous liquid pumping technique was successfully demonstrated for DI water and human blood.

**KEYWORDS:** Dielectrophoresis, Virtual microchannel, Particles trapping

## INTRODUCTION

In recent years, micro total analysis system ( $\mu$ -TAS) and lab-on-a-chip have been developed intensively. Among all the investigated technologies, pumping liquid in microchannels is essential to the study of microfluidics. There are different ways to drive liquid in microchannel, for example using hydraulic pressure difference along the microchannel or relying on the zeta potential on the microchannel wall. Previously, we reported a liquid pumping system in a virtual microchannel by dielectrophoresis (DEP) without physical microchannel walls [1]. Here we further develop a pumping technique to pump liquids in DEP-formed virtual channels continuously at tunable velocity between parallel plates by DEP.

## THEORY

The Maxwell stress caused by DEP generates a pressure difference across the liquid-medium interface expressed as [1, 2]:

$$\Delta P = P_M - P_L = \frac{\epsilon_0 (\epsilon_L - \epsilon_M)}{2} E^2 \quad (1)$$

where  $E$  is the electric field intensity,  $\epsilon_0$  ( $8.85 \times 10^{-12}$  F m<sup>-1</sup>) is the permittivity of the vacuum, and  $\epsilon_L$  and  $\epsilon_M$  are the relative permittivity of the liquid and medium, respectively. Recently, we have demonstrated drawing liquids of a higher permittivity as shown in Fig. 1(a) [1]. To pump the liquid continuously at a desirable flow rate, a constant pressure difference needs to be provided across a microchannel. Here two reservoir electrodes are designed adjacent to a microchannel electrode as shown in Fig. 1(b) and 1(c).

## EXPERIMENTAL

ITO (indium tin oxide) glass plates were used to be the substrates with defined electrodes by photolithography and wet etching. The top glass plate was covered by ITO and Teflon (55 nm-thick, DuPont AF 1600) layers. ITO electrodes were patterned on the bottom plate and covered by 1  $\mu$ m-thick dielectric (SU-8 2002, Micro Chem) and Teflon layers.

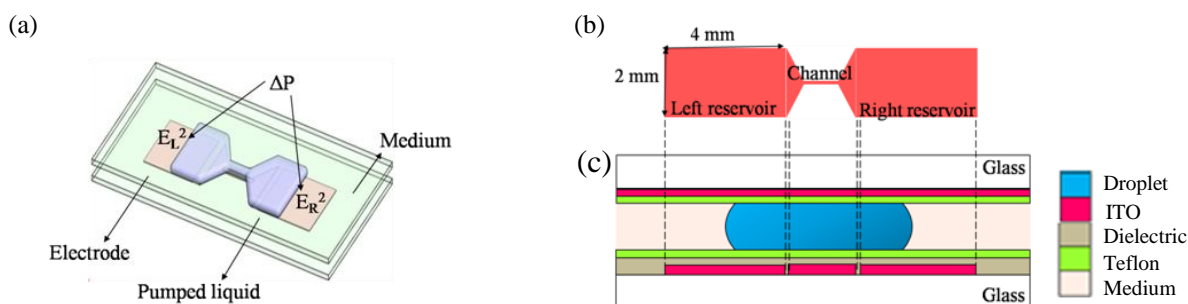


Figure 1: Configuration of the device. (a) Principle of pumping through pressure differences caused by DEP. (b) and (c) Top and cross-sectional views of the device, showing two reservoirs and one channel.

## RESULTS AND DISCUSSION

### Pumping DI water

0.4  $\mu\text{l}$  water ( $\epsilon_L = 80$ ) was dispensed in a 25  $\mu\text{m}$ -high gap between the parallel plates surrounded by 20 cSt silicone oil ( $\epsilon_M = 2.5$ ). Figure 2 shows water pumped from the right reservoir to the left one. After the virtual channel was formed by DEP when  $V_C$  was applied (Fig. 2(a)), a voltage higher than that applied on the right reservoir ( $V_R$ ) was applied on the left reservoir ( $V_L$ ) to continuously pump water in the 100  $\mu\text{m}$ -wide and 1 mm-long channel (Fig. 2(b)-(d))

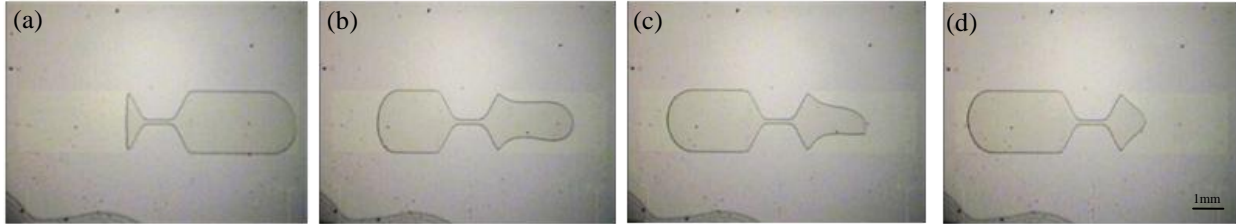


Figure 2: DI water pumping. 0.4  $\mu\text{l}$  water was pumped from the right reservoir to the left reservoir surrounded by 20 cSt silicone oil. The voltage supplied on the left reservoir ( $V_L$ ), channel ( $V_C$ ) and right reservoir ( $V_R$ ) were 57.6, 86.9 and 28.8  $V_{RMS}$ , respectively.

As shown in Fig. 3, when 60, and 35  $V_{rms}$  (100 kHz) were applied on the left reservoir and channel (150  $\mu\text{m}$ -wide) respectively. When different voltage as 80, 70, and 60  $V_{rms}$  were applied on the right reservoir, the flow rate in the microchannel was calculated as 96.3 nl/s, 69.7 nl/s, and 42.0 nl/s. From Eq. (1), the flow rate was proportional to the difference of  $E^2$ , which was in accordance with the results shown in Fig. 3. The ratio of the flow rates (1:1.67:2.29) was close to the ratio of  $\Delta E^2$  (1:1.55:2.18).

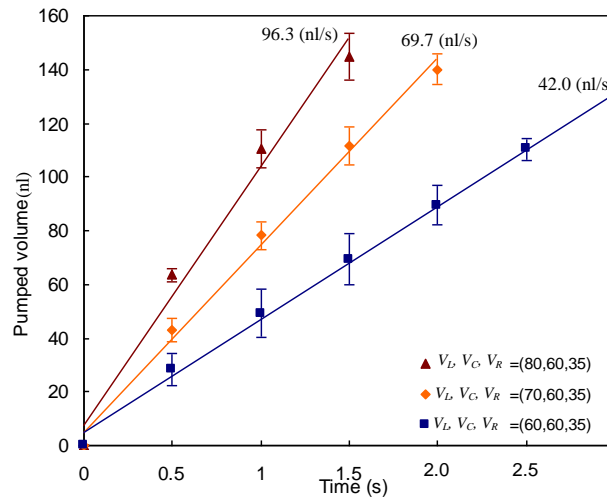


Figure 3: Continuous liquid pumping in a DEP-formed microchannel by applying different  $V_L$  and  $V_R$  to create various pressure gradients and flow rates. .

### Pumping human blood

Blood pumping in a 25  $\mu\text{m}$ -high gap between parallel plates surrounded by 20 cSt silicone oil was performed when applying the voltages. Biofouling of the blood was first severe, hindering the DEP driving and the pumping, so adding pluronic F127 as a surfactant could avoid the problem from the literature [3]. After adding pluronic F127 in blood, DEP was able to drive the blood by voltage application as shown in Fig. 4 (a)-(c).

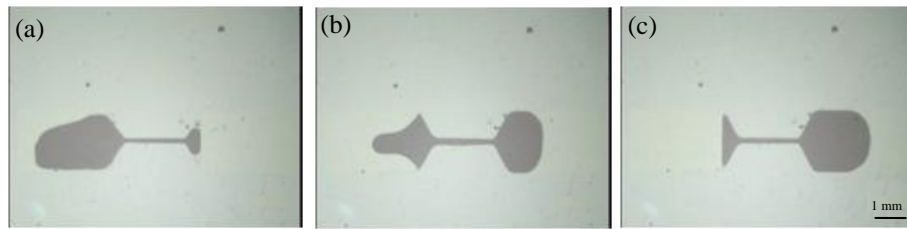


Figure 4: Pumping human blood. (a)-(c) When the voltages were applied, blood with pluronic F127 added was driven successfully.

#### Partical trapping

Properly designed microchannel electrode would establish non-uniform electric fields and manipulate the particles by DEP. As shown in Fig. 5 (a), the microchannel electrode between two reservoir electrodes was incomplete with cracks. The cracks provided weak and strong electric field regions. As shown in Fig. 5 (b), when a 100 kHz signal was applied, 5  $\mu\text{m}$  polystyrene beads were actuated by negative DEP and gathered above the cracks. Simultaneous particle and liquid manipulations were achieved.

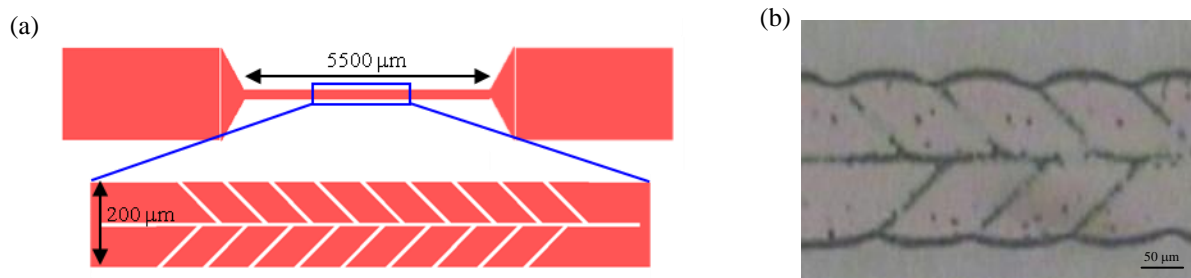


Figure 5: (a) The channel electrode design to create non-uniform electric fields, and the cracks were 10  $\mu\text{m}$ -wide. (b) 5  $\mu\text{m}$  polystyrene beads concentrating by negative DEP in the crack regions during liquid pumping.

#### CONCLUSION

In this research, the liquid pumping system with continuous flow rate was achieved. The flow rate was determined by the different voltage applied. By adding pluronic F127, human blood was successfully driven with eliminated biofouling problem on the chip. 5  $\mu\text{m}$  polystyrene beads were actuated by negative DEP and gathered above the cracks when a 100 kHz signal was applied. Therefore, simultaneous particle and liquid manipulations were achieved. The developed technique would be applied to blood cell separation during transportation.

#### REFERENCES

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