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Investigation of active interface control of pressure driven two-fluid flow in microchannels

Cheng Wang, Nam-Trung Nguyen, Teck Neng Wong, Zhigang Wu,
Chun Yang, Kim Tiow Ooi*

*School of Mechanical and Aerospace Engineering, Nanyang Technological University,
Nanyang Avenue 50, Singapore 639798, Singapore*

**Corresponding author. Tel.: +65 67905877; fax: +65 67911895.*

E-mail address: mtnwong@ntu.edu.sg.

Abstract

We report a novel concept to control the interface location of a pressure-driven multi-phase flow in a microchannel by using electroosmotic flow effects. This concept has potential applications in flow switching and cell sorting in bio-analytical systems. In an H-shaped microchannel structure, aqueous sodium chloride (NaCl) solution and glycerol diluted with water were pumped through two inlets at the same flow rate. The electric field was applied on the electrolyte solution side. Adjusting the magnitude and direction of electric field has successfully controlled the interface position between the two phases. This technique provides a new approach to control the interface position between the two fluids.

Keywords: Active interface control; Two-fluid flow; Electroosmosis; Microfluidics

1. Introduction

One fundamental characteristic of flows in microfluidics is the low Reynolds number. The consequence is that the microflow is in the laminar flow regime in most microfluidic systems [1]. Two or more streams co-flowing in a microchannel will flow side by side, without convective mixing, except diffusion at the interface where they are in contact. The interfaces between the multiple flow streams are called the laminar fluid diffusion interfaces (LFDIs) [2].

Diffusion-based microfluidic devices, such as T-sensor and H-filter [3–5], are based on the LFDIs. For a typical T-sensor, two streams are brought to form a co-flow in the main channel. Optical detection is performed at the diffusion interface where chemical reaction occurs. The H-filter is used for separation and extraction purposes. The H-shape has two inlets and two outlets. Two streams, i.e., sample fluid and buffer fluid are introduced through the two inlets. The sample contains two types of particles or cells to be separated. In the main channel, diffusion occurs at the interface. Big particles have small diffusivity and diffuse much slower than the small particles, and stay in the sample stream. At the end of the main channel, small particles or cells will diffuse to the buffer solution and are separated.

The variable properties of biological samples can be problematic when the two streams have different viscosities [6]. The fluid with higher viscosity will spread wider in the channel while having a lower velocity; whereas the fluid with lower viscosity flows at a higher velocity occupying a smaller fraction of the channel volume. The interface position deviates from the central line position. The above observation applies when the two flows have the same volumetric flow rate. The unmatched viscosity affects the diffusion and reaction kinetics in terms of the residence time. The average residence time of the more viscous fluid increases, while that of the less viscous fluid decreases.

In the literature, the two conventional methods applied to solve the variable viscosity problem are using viscosity enhancing additives to match the viscosities of the two fluids and adjusting the flow rate ratio [7]. In the first method, viscosity has to be measured first and precise viscosity enhancing material control is required. The second method can shift the interface position to the center of the main channel, but still cannot solve the problem of unmatched residence time, because the fluid with higher flow rate still flows faster. In this paper, a novel concept to control the interface position using electroosmosis (EOF) is presented. Parameters, such as flow rate and electric field, which affect the interface position of the two-fluid flow are investigated.

2. Interface control concept

An H-shaped microfluidic device is used for demonstrating our concept of interface control, in Fig. 1. Two fluids, aqueous NaCl solution and diluted glycerol are introduced

through the inlets by a syringe pump. The volumetric flow rates for NaCl solution and diluted glycerol are maintained at the same value. Electric field is applied on the NaCl solution flow side. An electroosmosis flow results for the NaCl solution flow side.

For a purely pressure-driven two-fluid flow, the aqueous NaCl solution with a lower viscosity flows faster and occupies a smaller width of the channel. Thus, the LFDI position is not at the channel center. NaCl solution also has a smaller residence time because of the higher velocity. The working principle of interface control using EOF is depicted in Fig. 2. Once an electric field is applied, the velocity of NaCl solution flow can be adjusted. If the EOF is of the opposite direction of the pressure-driven flow, the NaCl solution will be slowed down. Consequently, the interface position is shifted to the central line. The interface position moves further away from the central position if the EOF is in the same direction of the pressure-driven flow, because of the increase of the NaCl solution velocity.

3. Experimental setup

The H-shaped microfluidic device used in our experiment was based on the adhesive lamination technique, shown in Fig. 3. In this method, two polymethylmethacrylate (PMMA) plates (75mm x 25mm) are bonded by one layer or many layers of double-sided adhesive tapes to form a closed microfluidic channel with inlet and outlet holes. The channel structure was cut through the adhesive tape using a CO₂ laser cutting system (Universal Laser Systems Inc., laser power of 25 W, and a maximum beam speed of about 640 mm/s). Thus, the adhesive tapes define the depth of the channel. In this experiment, the straight portion of the H-shaped channel has the cross section of 910 μm x 100 μm and the length of 5 cm. Platinum electrodes were inserted for the application of the electric field.

A schematic of the experimental measurement setup is shown in Fig. 4. The setup consists of four main components: an illumination system, an optical system, a coupled charger device (CCD) camera and a control system. The control system consisting of a peripheral component interface (PCI) card, and its corresponding software, is implemented in a personal computer (PC). The PC can control and synchronize all actions related to illumination and image recording. The NaCl solution and diluted glycerol (volume concentration of 14%) were introduced through inlets A and C, as in Fig. 1. The schematic cross sectional view of the two fluids flow inside the straight channel is shown in Fig. 5. The

width to depth ratio used in this experiment is 9.1:1. The widths occupied by the NaCl solution and diluted glycerol are denoted as w_2 and w_1 , respectively. The holdup of the NaCl, e_2 , is defined as the ratio of the area occupied by the NaCl to the whole area of the cross section of the channel, given as $e_2 = w_2/(w_1 + w_2)$. Similarly, the holdup of the diluted glycerol is $e_1 = 1 - e_2$.

Fluorescent dye (fluorescein disodium salt $C_{20}H_{10}Na_2O_5$, also called Acid Yellow 73) was added in the NaCl solution for images collection. When the fluorescein was illuminated by a mercury lamp, CCD camera (Nikon B-2A) was used to take the intensity image. The same volumetric flow rates of the two inlet flows were ensured through two identical syringes driven by a single syringe pump (Cole-Parmer, 74900-05, 0.2–500 mL/h, accuracy of 0.5%). After recording the images, the interface location between the two fluids was determined by an algorithm written in a MATLAB program [8]. The algorithm removes the noise in the collected images, and then determines the interface position between the two fluids. A proportional relationship is assumed between the fluorescent intensity and the concentration of the dye [8]. The interface position between the two fluids can be determined from the evaluation of the concentration gradient. The typical images taken in the experiment and the concentration profiles are shown in Fig. 6.

4. Results and discussion

The parameters studied in the experiments are the inlets volumetric flow rates, the electric voltage applied between A and B and the viscosity ratio β . Viscosity ratio β is defined as the ratio between the viscosity of the diluted glycerol solution and that of the NaCl solution, given as $\beta = \mu_1/\mu_2$. For different operating conditions, the measurement was taken when the flow is in steady-state condition. For results in Sections 4.1 and 4.2, the viscosity ratio is kept constant at $\beta = 1.5$.

4.1. Effect of applied voltage

As shown in Fig. 7, as the electric field varies in magnitude and direction, the holdup of the NaCl solution changes accordingly. When no voltage is applied across A and B, the flow is simply a pressure-driven two-phase flow. Since the glycerol solution is more viscous than the NaCl solution, $\beta = 1.5$, the NaCl solution occupies a smaller portion of the channel, and has a

holdup of 0.35, as shown in Fig. 7. When a negative electric field is applied on the NaCl solution side, the electroosmotic flow is against the pressure-driven flow. Under a negative electric field, more resistance force will be encountered to drive the NaCl solution. Apparently, the NaCl solution becomes more “viscous” due to the additional electroosmotic force. Thus, the NaCl solution occupies a larger channel width, i.e., the holdup of the NaCl solution increases. When a positive electric field is applied, the NaCl solution has apparently a lower “viscosity”, since the electroosmotic force acts in the same direction as the pressure gradient. In another words, electroosmotic effect aids the flow of the NaCl solution.

4.2. Effect of flow rate

The relationship between the NaCl holdup e_2 at different flow rates under a constant electric field is shown in Fig. 8. Holdup $e_2 = 0.35$ remains the same for different volumetric flow rates in the absence of an externally applied electric field, because the flow rates ratio between the two fluids is kept constant at 1:1. This result agrees well with a previous analytical and numerical study reported in literature [7]. Fig. 8 shows that holdup e_2 converges to a constant value of 0.35, as flow rate increases. This value is the holdup without the external applied electric field. When flow rate is higher, the flow induced by a pressure gradient dominates over the electroosmosis flow. Since the electroosmotic force is negligible compared to the pressure gradient, the flow in such cases can be considered as a pure pressure driven flow.

4.3. Effect of viscosity ratio

To investigate the effect of viscosity ratio, diluted glycerol solution of two different viscosities, were used in the experiment. The viscosity ratios are $\beta = 1.5$ and 2, respectively. The interface positions for different β values are shown in Fig. 9. The channel in the experiment has the cross section of $910 \mu\text{m} \times 100 \mu\text{m}$. Fig. 9(a) shows that for the case of pressure driven two-fluid flow, i.e., in the absence of applied electric field, the holdup of the NaCl solution is about $e_2 = 0.35$ and 0.31 for $\beta = 1.5$ and 2, respectively. The holdup of the NaCl solution decreases with the increase of the viscosity ratio, because the more viscous diluted glycerol solution spreads to a larger width. Furthermore, we can observe that the gradient of $\beta = 1.5$ is steeper than that of $\beta = 2$.

For a given applied electric field, the difference between the “apparent viscosity” μ_{app} and the actual viscosity is $\Delta\mu_2 = \mu_{\text{app}} - \mu_2$. The relative change of viscosity μ_2 with respect to

the fluid 1, i.e., glycerol, is $\Delta\mu_2/\mu_1 = \Delta\mu_2/\beta\mu_2$. Thus, for a larger β value the relative change of μ_2 with respect to μ_1 is smaller. The interface position can be adjusted more effectively for the two fluids when they have comparable viscosities, i.e., β close to 1. At the flow rate of 0.8 mL/h, there seems no holdup change, though the applied voltage varies from 0 till -0.8 kV for $\beta = 2$. But for $\beta = 1.5$, the holdup is still controllable at the flow rate 0.8 mL/h. That means, for viscosity ratio close to 1, the electroosmosis effect is most effective to control the interface position.

5. Conclusions

Our paper reports a novel way to control the interface position between two pressure driven streams in a microchannel. The interface position of two-fluid flow in microchannel can be controlled using the electroosmosis effect, rather than the conventional “flow-rate-ratio” method. The method uses electroosmosis caused by a control voltage to adjust the interface position. This method is effective for low flow rates, and small viscosity ratio close to 1. The low flow rates and small viscosity ratio make sure that the pressure gradient and electroosmotic force are of the comparable order. The experiment demonstrated a new method to solve the unmatched viscosity problem of the two-fluid flow in microchannel, which are very often used in biological extraction, separation and sorting processes.

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Biographies

Cheng Wang received his BE (first class honors) degree in mechanical engineering from the Nanyang Technological University, Singapore in 2004. He studied for the Master of Engineering degree in the School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore from 2004 to 2006, in the area of the two-fluid flow in microchannel for diffusion-based biomedical applications. He is now working in the Micromachines Center, School of Mechanical and Aerospace Engineering of Nanyang Technological University as a research staff, with focus on the fabrication of Photonic Crystals.

Nam-Trung Nguyen was born in Hanoi, Vietnam, in 1970. He received his Dip -Ing, Dr Ing and Dr Ing Habil degrees from Chemnitz University of Technology, Germany, in 1993, 1997, and 2004, respectively. In 1998, he worked as a postdoctoral research engineer in the Berkeley Sensor and Actuator Center (UC Berkeley, USA). Currently, he is an Associate Professor with the School of Mechanical and Aerospace Engineering of the Nanyang Technological University in Singapore. His research is focused on microfluidics and instrumentation for biomedical applications. He published a number of research papers on microfluidics. His recent book “Fundamentals and Applications of Microfluidics” co-authored with S. Wereley was published in October 2002. The second edition of this book with updated contents was published in 2006.

Teck Neng Wong received his PhD in mechanical engineering from University of Strathclyde, UK in 1990. He is an associate professor in School of Mechanical and Aerospace Engineering, Nanyang Technological University. His research interests include two phase flow and heat transfer, flow boiling phenomena, two-phase flow in evaporator, condenser and capillary tube expansion devices, heat driven pump and pulsating heat pipe for electronic cooling; micro-scale heat transfer and fluid flow, two-fluid electroosmotic pump, electroosmotic control of the interface between two fluids in microchannels and droplet transport in microfluidic devices.

Zhigang Wu was born in Hubei, China. He received his BSc degree from Huazhong University of Science and Technology (HUST), China in 2001 and PhD degree from Nanyang Technological University (NTU), Singapore in 2005, respectively. He is currently a research fellow working on microfluidics at the School of Electrical and Electronic Engineering of NTU. His research interests include: polymeric lab-on-a-chip devices and system for clinical application and home healthcare; rapid microfabrication techniques for lab-on-a-chip applications, BioMEMS components and their integration; transport effects in microscale and their applications in life-sciences and chemistry.

Chun Yang obtained his BSc from Tsinghua University in 1985 and PhD from University of Alberta in 1999. He was with Syncrude Research Canada under NSERC Oil-Sands Research Chair Program. Now he is an associate professor in School of Mechanical and Aerospace Engineering, Nanyang Technological University. His current research interests include microfluidics, electrokinetic and interfacial transport phenomena, colloidal particle and droplet transport and interaction, complex fluid mechanics and nanofluids heat transfer.

Kim Tiow Ooi received his BE and PhD in Mechanical Engineering both from University of Strathclyde, UK in 1986 and 1989, respectively. He is an associate professor in School of Mechanical and Aerospace Engineering, Nanyang Technological University. His current research interests are in refrigeration devices, heat transfer, and fluid flow in macro and micro devices. He is widely consulted on design, optimization and fabrication of refrigeration compressors both locally and internationally.

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Fig. 1

(a) Pressure driven only

(b) Negative electric field

(c) Positive electric field

Fig. 2

Fig. 3

Fig. 4

Fig. 5

Fig. 6

Fig. 7

Fig. 8

Fig. 9