STABILITY OF PRESSURE-DRIVEN FLOW IN OPEN FLUIDIC CHANNELS

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Summary Microfluidics involves the study of behavior and manipulation of fluids at sub-millimeter scales [1]. This work is focused on open microfluidic systems where fluid is transported by means of pressure-driven flow. Experiments were conducted for straight open microchannels where variables include channel lengths, fluid flow rates, and wetted volumes. By identifying the limit of stability of such open microfluidic channels, changes can be made to the device promoting scalability and versatility of such systems.

INTRODUCTION

Microfluidic devices revolutionize industries of today including clinical and forensic analysis; microreaction engineering; surface patterning; and optical engineering. These devices allow fluid manipulation with fast response time, enable the handling of small fluidic volumes, and offer the advantage of sensing and controlling flow on small length scales. Open microfluidic systems are characterized by the presence of free-liquid-vapour or liquid-liquid interfaces. Typically these devices involve a single planar platform where fluid can be deposited and manipulated. Pressure driven flow is derived when there is difference in pressure between the beginning and the end of a microchannel through which liquid flows.

EXPERIMENTAL METHOD

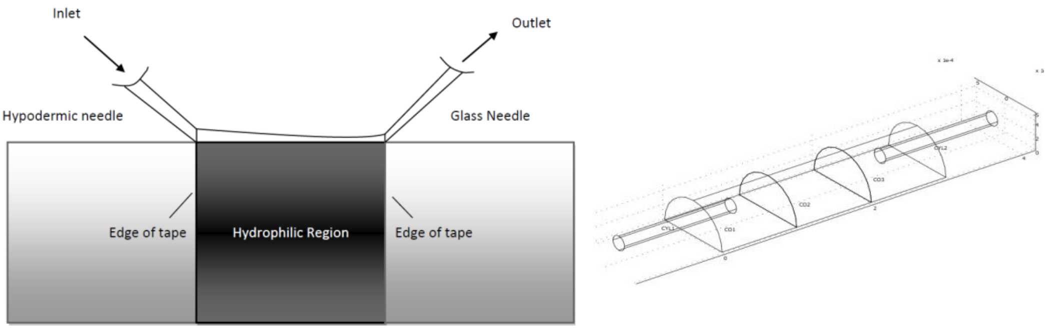

Figure 1: (Left) Schematic description of the experimental setup used to investigate pressure driven flow across the channel. (Right) Computational model showing fluid being infused and extracted via needles generated through COMSOL Multiphysics 3.3a.

The experimental setup is depicted in Figure 1where a glass microscope slide was positioned in a way that the long side edge was horizontal providing a surface channel width of approximately 1 mm. The testing region of the glass microscope, labeled as the hydrophilic region was pre-wetted with de-ionized water and the tubes from the syringes to the needles were pre-bled in advance. To distinguish the area outside the testing region, the left and right hand side of the testing region were wounded by tape. Channel length can then be changed accordingly as the hydrophilic region is limited by tape. Two needles were then attached to a syringe pump providing steady controllable flow rates. Flow rates were varied accordingly using this pump which allowed the channel to grow or reduce in height. The hypodermic needle was used as the input needle as it was hydrophobic in nature and assisted in pumping fluid into the system. The glass needle on the other hand was used as the output needle as it was more hydrophilic and aided in extracting fluid out of the microchannel. The resulting fluid flow profiles were captured using a CCD camera (Hitachi, KP-D20AU) attached to a telescopic lens (InfiniVar Video Microscope). Experiments examined a range of flow rates from slow (50 µl/min), medium (350 µl/min) and fast (650 µl/min) along with different channel lengths from short (5 mm), medium (15 mm) and long (30 mm). The influence of the initial wetted volume was also investigated.
Experimental results were compared to results obtained through computational models generated using COMSOL Multiphysics 3.3a. Models were designed using a constant height of 0.5 mm and constant width of 1 mm. The finite element model used a volumetric input consistent with the flow required and a zero pressure outlet. The model was capable of predicting maximum flow velocities and pressure head across the channel length.

RESULTS

An aspect which must be taken into account when varying fluid flow rates is the stability of the channel when flow is increased. When fluid profile stability is achieved for a given flow rate, increasing the flow rate will cause an increase in pressure head. The fluid profile must then readjust to compensate for this change. Once the channel height becomes half the width of the channel, any further growth in height will cause the radius of curvature to decrease and thus

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becoming a limiting factor. At high initial volumes where the fluid is heavily pre-wetted; a bulge develops and once the bulge exceeds the limiting contact angle associated with the edge of the channel; the stability of the microfluidic channel is compromised.

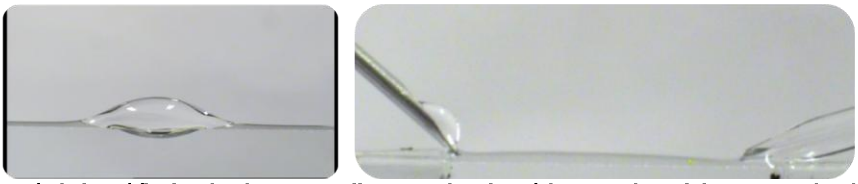

Figure 2: (Left) Formation of a bulge of fluid with subsequent spillage over the edge of the microchannel demonstrates breakdown in the microfluidic system. (Right) When the fluid layer downstream becomes very thin, the extraction needle becomes less efficient at removing fluid, a pulsing effect can then be observed beneath the needle.

The exposed nature of the open microfluidic system allows for external interaction. It is reasonable to deduce that the volume of fluid in the channel at any given moment of the experiment is the volume of fluid used to pre-wet the channel length due to the consistency offered by the syringe pump. The initial pre-wetting volume thus becomes the limiting factor to the stability of the channel. From previous computational efforts, it is observed that the optimum wetting volume for a microfluidic channel at any point of time is for the channel height to be half the width of the channel. From the experiments conducted, for a system with medium channel length (15 mm) and low flow rate (50 µl/min), the largest volume of droplet which can be dispensed into the system before breakdown of the channel occurs is 12.5 µl. Since this volume is fairly large in comparison to the flow rate, it can be deduced that small volumes of fluids can be added into the system without causing instability.

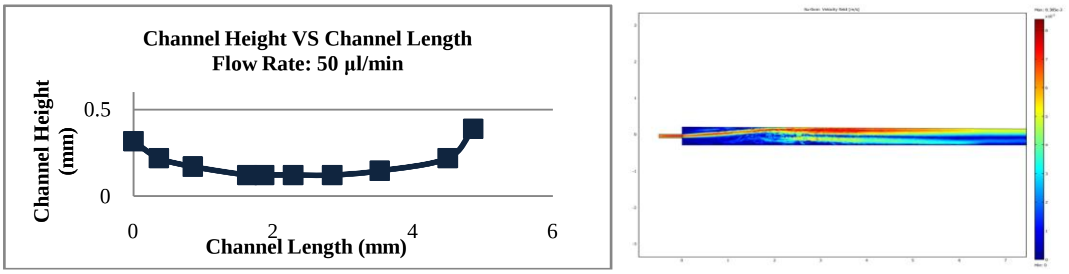

Figure 3: (Left) Curved shaped profile for a system with short channel length (5 mm) and low flow rate (50 µl/min). (Right) Result generated through the computational model for the side profile of fluid velocity flowing into the system. The effect due to the infusion needle can be observed where flow appears to be mixed and in transition before laminar flow is achieved further along the microfluidic channel. Velocity is also observed to surge at the inlet of the channel.

The stability of the fluidic profile is also influenced by the length of the microfluidic channel. For the short microchannel (5 mm), a curved shaped profile with raised ends is observed in Figure 3. This effect is deemed to be contributed by placement of needles at both ends. Placement of the infusion needle at the inlet angle of about 45 degrees to the channel creates an outward push of fluid into the microfluidic system. This effect is apparent in short profiles as fluid is not permitted to redistribute evenly before reaching the extraction end. Previous study by Davey (2011) [2] has also highlighted the influence of channel length in the stability and profile of the flow. Shorter channels were found to be more prone to display a bulge and the distortion due to the presence of needles was more prevalent.

CONCLUSIONS

Through the experiments conducted, pressure driven flow was investigated in open microfluidic systems. Breakdown of the microfluidic device occurred due to the compromised stability of the channel. When varying flow rates of fluid into the channel, the profile of the fluid was observed to be able to adjust to the pressure head which occurred over a constant channel length. Stability of the system is contributed to when limit in allowable pressure head is exceeded causing spillage of a bulge of fluid or the extraction needle unable to remove fluid effectively. For the case of adding droplets into the system, the system was observed to be able to adapt to irregular input of fluid volume. Adding small volumes of fluid into the flowing channel has been demonstrated without causing instability. The stability of the system however, is limited by the contact angle of the surface. By varying channel length, it can be concluded that the length of the channel plays a role in both fluid profile and stability of the flow. Stability of the fluid profiles was found to be influenced by the momentum contributed by placement of the needles.

References

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[2] Davey, N. and A. Neild, *Pressure-driven flow in open fluidic channels.* Journal of Colloid and Interface Science, 2011. **357**(2): p. 534-540.