LAB ON A CHIP
Communication

A passive pumping method for microfluidic devices

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The surface energy present in a small drop of liquid is used to pump the liquid through a microchannel. The flow rate is determined by the volume of the drop present on the pumping port of the microchannel. A flow rate of 1.25 μL s^{-1} is demonstrated using 0.5 μL drops of water. Two other fluid manipulations are demonstrated using the passive pumping method: pumping liquid to a higher gravitational potential energy and creating a plug within a microchannel.

Introduction

An efficient and easy to implement pumping method is mandatory for making microfluidic devices a ubiquitous commodity. Several non-traditional pumping methods have been developed with some promising results.[1–4] However, the one drawback almost all pumping methods have is the requirement for expensive or complicated external equipment, be it the actual pumping mechanism (*e.g.*, syringe pumps), or the energy to drive the pumping mechanism (*e.g.*, power amplifiers). The ideal microscale pump would be semi-autonomous and incorporated totally at the microscale—the biomimetic equivalent of an insect's circulatory system[5] or blood-sucking apparatus.[6] Here we report a simple method for pumping fluids that is semi-autonomous and only requires a device capable of producing small drops of liquid, such as a pipette. Because a pipette is all that is needed, this pumping method allows microfluidic devices to be used in a variety of settings—from small labs to industrial high-throughput robotic assaying systems.

Theory

The amount of pressure present within a drop of liquid at an air/liquid interface is given by the Young–LaPlace equation $\Delta P = \gamma(1/R_1 + 1/R_2)$, where ΔP is the difference between atmospheric pressure and the pressure inside the drop, γ is the surface free energy of the liquid, and R_1 and R_2 are the radii of curvature for two axes normal to each other that describe the curvature of the surface. For spherical drops, $R_1 = R_2$, and this equation reduces to $\Delta P = 2\gamma/R$, where R is the radius of the sphere. A consequence of the Young–LaPlace equation is that smaller drops have a higher internal pressure than larger drops. Therefore, if two drops of different size are connected *via* a fluid-filled tube, the smaller drop will shrink while the larger one grows. One manifestation of this effect was observed in a capillary electrophoresis device when a flow was induced by unequal reservoir fluid levels.[7] The proposed passive pumping method harnesses this phenomenon to perform useful work in microfluidic systems.

Fluid can be pumped through a microchannel by using the surface tension in a drop of liquid and two or more ports on the microchannel—a reservoir port and a pumping port. A large drop (*e.g.*, 100 μL) is placed over the reservoir port of a fluid-filled microchannel. The radius of this drop is large enough to cause the pressure at this port to essentially be zero. A much smaller drop (*e.g.*, 0.5–5 μL) is placed on the pumping port, and because of its smaller radius, a larger pressure exists at this end of the microchannel (Fig. 1a). The Young–LaPlace pressure in the reservoir drop of height L is approximately 0 Pa and, assuming a hemispherical drop of radius 0.06 cm, the pressure at the pumping port is 243 Pa (the maximum pressure attainable for this port size). The resulting pressure gradient causes fluid to flow through the microchannel towards the reservoir drop. The highest pressure attainable for a given pumping port radius is a hemispherical drop whose radius equals that of the port. Any deviation from this size, either larger or smaller, results in a lower pressure.

The radius of the pumping drop can be found by first solving for the height, h, that the drop rises above the port (Fig. 1b). The height of a spherical cap of volume V can be found using the equation

$$h = \frac{1}{6}[108b + 12(12a^3 + 81b^2)^{\frac{1}{2}}]^{\frac{1}{3}} - \frac{2a}{[108b + 12(12a^3 + 81b^2)^{\frac{1}{2}}]^{\frac{1}{3}}} \quad (1)$$

where $a = 3r^2$ (r is the radius of the port cored into the microchannel) and $b = 6V/\pi$ (V is the volume of the drop placed on the port). The drop radius, R, can then be found by

$$R = \left[\frac{3V}{\pi} + h^3\right]\frac{1}{3h^2} \quad (2)$$

The volumetric flow rate from the pumping port to the reservoir port will change as the drop volume changes. The change in volume with respect to time can be calculated using the equation

$$\frac{dV}{dt} = \frac{1}{Z}\left(\rho g L - \frac{2\gamma}{R}\right) \quad (3)$$

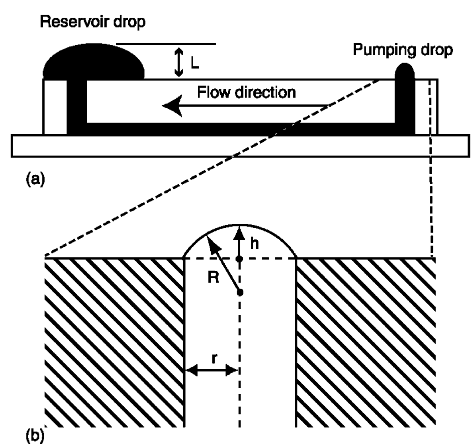

Fig. 1 (a) Side view of a microchannel. A reservoir port with a large drop and pumping port with a smaller drop are required for fluid flow. (b) A drop of volume V will form a spherical cap of radius R on a port of radius r. The cap will rise above the surface of the device a distance h. If the drop volume is less than that for a hemisphere of radius r, then the drop radius, R, will be larger than h.

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where R is the radius calculated in eqn. 2 and Z is the resistance of the microchannel. The pressure created by the reservoir drop of density ρ and height L is calculated by ρgL (Fig. 1a). The height of the reservoir drop is constant above a certain volume. Because of its large volume, the pressure created by the reservoir drop is due to gravity and not surface tension effects.

Experimental

Microchannels were constructed using polydimethylsiloxane (PDMS) and rapid prototyping.[8] Ports were cored down to the microchannels using a blunt 16 gauge syringe needle (id 1.2 mm). PDMS was chosen partly because of its extreme hydrophobicity, which prevented drops of fluid on the pumping port from spreading out and reducing the Young–LaPlace pressure.

The calculation of eqn. 3 was done with MATLAB using the fourth-order Runge–Kutta method. The observed results were obtained by measuring the height of one pumping drop every 0.067 s until the height became constant (*i.e.*, until the pumping drop volume was completely exhausted). Using this method, three consecutive drops were observed and the heights at each time increment averaged together. The corresponding volume for each average height was then calculated and plotted with the predicted volumes in Fig. 2a.

This pumping method was also used to move water to a position of higher gravitational potential energy. If the reservoir drop is placed at a height greater than the pumping drop, the result will be water that pumps itself uphill. To demonstrate, a microchannel was constructed in which the reservoir port was elevated above the pumping port with two pieces of PDMS, creating a port 0.8 cm taller than the pumping port (Fig. 3a). Both PDMS pieces had been cored out with the 16 gauge needle. After the microchannel and both ports were filled with water, a large drop was added to the reservoir port and a 0.5 μL drop of blue dye was added to the pumping port. Once the blue dye was mixed thoroughly in the port, 1 μL drops of water were added until the majority of the dye had been moved from the pumping port to the reservoir port (Fig. 3b and 3c)—a total of 26 drops.

The height to which water can be pumped is limited by the pressure generated from the height of the water column within the reservoir port. The water column pressure effects can be seen on the pumping port (Fig. 3a, inset). At equilibrium, the pumping drop assumes a slightly curved shape compared to Fig. 2d, in which the reservoir and pumping ports are at the same height.

To demonstrate applicability to a common microfluidic task, this pumping method was used to move a packet of fluid down a microchannel. The device used was a t-junction in which all microchannels were initially filled with water. One microchannel was then filled with blue dye using the passive pumping method (Fig. 4a). A 0.5 μL drop of water was added to the port on the right. The effect was to move a packet of blue dye out of its stream a distance of 4.5 mm while simultaneously pushing blue dye back into each side microchannel (Fig. 4b).

Results and discussion

One prediction of the model, and confirmed by observation, is the rate of change in drop volume is fairly constant for half of the pumping period. The small standard deviations of the observed drop volumes indicate this pumping method provides repeatable pressure-driven flows and essentially constant flows for the first half of the drop volume. In contrast, syringe pumps are unsteady at very slow flow rates and can vary in pressure by as much as 50% of the mean pressure value.[9]

Once the pumping drop volume is less than half of its initial value, the volumetric rate of change with respect to time becomes nonlinear. In this region of Fig. 2a, the observed volumes at $t = 0.2$ and 0.267 s are slightly greater than the predicted values. This discrepancy can be attributed to the sensitivity of the model to quantities which were not accurately known, such as the surface free energy of the blue dye, the pumping drop volume, and the microchannel resistance.

The pumping rates achievable using this method depend on the volume of the drop, the channel resistance, and maintaining the pumping drop shape. For the device in Fig. 2, it took 0.4 s for a 0.5 μL drop to deplete itself—a pumping rate of 1.25 μL

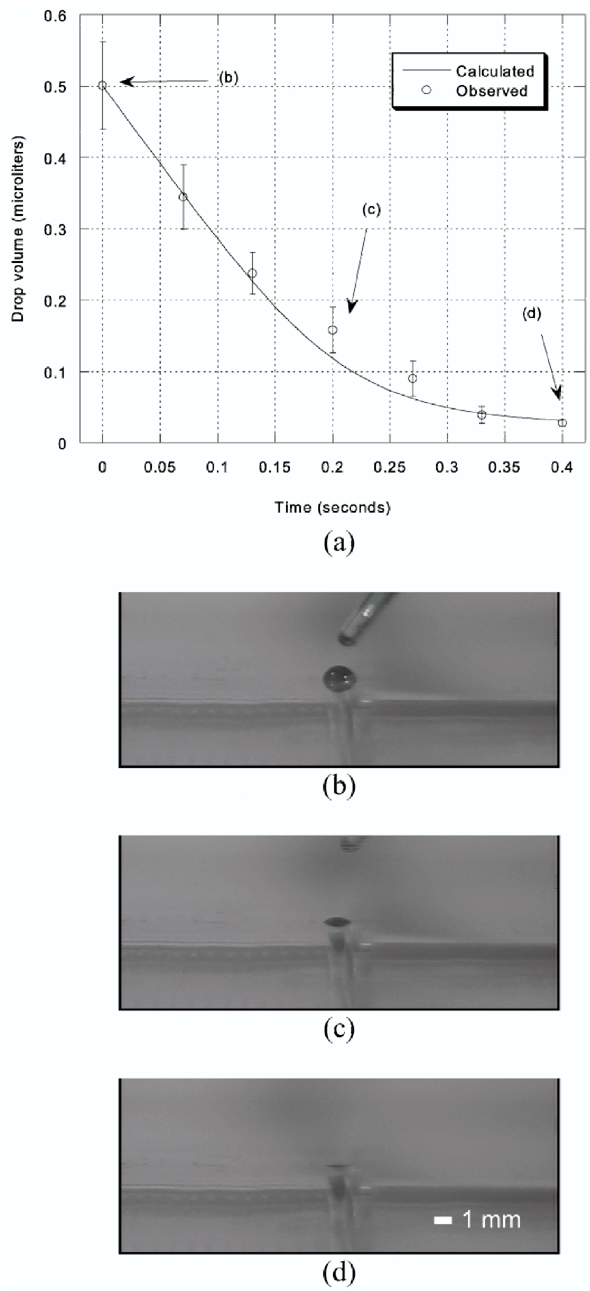

Fig. 2 The height of a single drop on the pumping port was measured from pictures taken at 0.067 second intervals during the pumping process and converted into a volume. The observed average volumes of three drops are plotted against the calculations from the model (a). Microchannel dimensions were 2 cm (L) × 140 μm (H) × 1 mm (W). The ports had a radius of 0.06 cm. (b–d) View of the pumping port at $t = 0$, 0.2, and 0.4 s, respectively. The drop on the pumping port can be seen going down as fluid travels down the microchannel.

s^{-1}. As mentioned before, the pumping drop volume change with respect to time eventually becomes nonlinear. If pumping drops were supplied at a rapid enough rate, before the nonlinear region was reached, the change in volume with respect to time would resemble a sawtooth waveform. Even though the volume *versus* time waveform would not be constant, the fluid flow within the channel would be steady with occasional disturbances from the addition of extra pumping drops. As an analogy, imagine a macroscale pump of constant flow rate with a finite input reservoir of fluid. If fluid is added every so often to keep the reservoir full, the pumping rate will not change. However, a plot of the change in input reservoir volume *versus* time would resemble a sawtooth waveform.

The channel resistances are another factor which affect the pumping rate. Although the channels used in the experiments reported here had larger dimensions by microscale standards, the pumping method would work just as well with much smaller dimensions. The effect of smaller channels, and thus greater flow resistances, is simply to stretch the volume *versus* time graph in the time dimension. This result follows from eqn. 3, since the derivative is inversely proportional to the channel resistance.

The third aspect which affects pumping rate is the shape of the pumping drop. Minimizing the surface area of the pumping drop is important for this pumping method to work well. As mentioned previously, PDMS was used because its hydrophobicity kept the pumping drop beaded, thus maximizing the Young–LaPlace pressure within the drop. More hydrophilic materials could be used for device construction, but they would cause the pumping drop to spread, thus increasing the drop radius and reducing the amount of pressure available for pumping. For example, if this method were used with a glass device, it probably would not work due to the extremely low contact angle of water on glass.

Even though evaporation occurs while using this pumping technique, it is often not a problem for several reasons. First, the pumping drops displace fluid in the microchannels at a rapid enough rate to render evaporation-induced flows negligible. Second, if slower rates or smaller drops are desired, the pumping can either be done in a humidified local environment (*e.g.*, in a petri dish with water in the bottom) or the drop volume can be increased to compensate for the evaporation losses. Third, if this method were to be used over an extended period of time, then pumping drops composed of dilute solutions could be used to maintain constant concentrations of particles in the microchannels.

Other applications of this pumping method exist. For example, multiple ports could be cored along the length of a microchannel. By designating one the reservoir port, different flow rates could be achieved by pumping from different pumping ports along the microchannel length (due to the difference in microchannel resistance). Also, temporary reservoir ports could be used to cause fluid to flow into them, mix, and then in turn be pumped to other reservoir ports. In addition to water, this pumping method works with biological fluids. Media containing cells and fetal bovine serum was used to successfully and repeatedly flow cells down a microchannel.

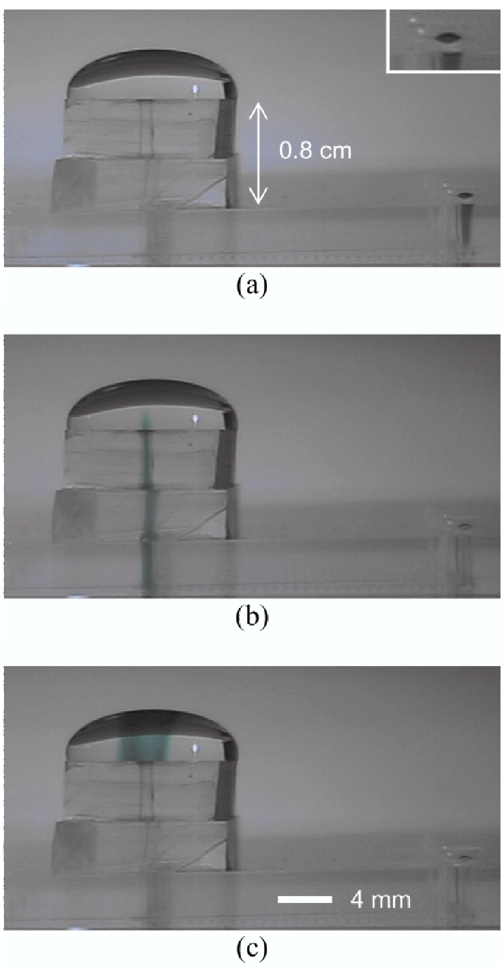

Fig. 3 (a) Using the same microchannel dimensions as in Fig. 2, a new device was made in which the reservoir port was elevated above the pumping port by 0.8 cm. A 0.5 μL drop of blue dye was then added to the pumping port. The increase in reservoir port height, and therefore pressure in the water column, caused the resting drop at the pumping port to bulge up slightly (a, inset). (b) Seven drops of water later, most of the blue dye had been moved into the left vertical channel (id 0.12 cm). After a total of 26 water drops, the majority of the blue dye had been transported to the reservoir drop (c).

Conclusion

The most important benefit of this pumping method is that it can be implemented in a wide variety of settings. For example, it can

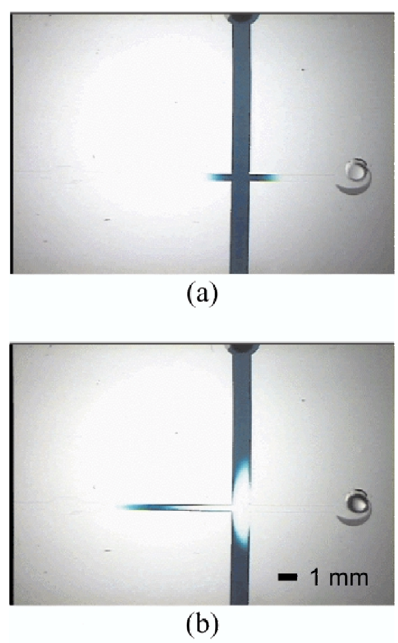

Fig. 4 After the entire device was filled with water, the microchannel with blue dye was filled using passive pumping. Its dimensions were 50 μm (*H*) × 1 mm (*W*) × 2 cm (*L*). Once filled with blue dye, the reservoir drop was removed and another one was placed on the port to the left (not shown). A water drop was then placed on the port to the right. This pushed a plug out of the blue microchannel while simultaneously pushing the blue back into its microchannel on each side (b). The horizontal microchannel dimensions were 50 μm (*H*) × 400 μm (*W*) × 3 cm (*L*).

easily be interfaced with high-throughput robotic systems or used in laboratories where much simpler equipment is available. The only equipment needed is a device which can deliver small quantities of liquid, such as a syringe or pipette.

Because this method relies on surface tension effects, it is repeatable and can be tailored to accommodate many different pumping applications. The pumping rate can be adjusted by changing any combination of factors: the volume of the pumping drop, the surface free energy of the liquid, or the resistance of the microchannel, to name a few. All these factors are controllable to a high degree of precision with modern laboratory equipment and allow for a pumping method suitable for use in a variety of situations and applications.

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