A compact syringe-assisted vacuum-driven micropumping with a constant flow rate

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ABSTRACT

In this paper, a simple syringe-assisted pumping method was introduced. Compared with a previous study [1], the proposed design does not require pre-designed vacuum chamber. Using proposed pumps, multiple pumps can be integrated into one device as well as a constant flow rate can be achieved. This method enables a point-of-care pumping actuation methodology with more flexibility in hand-held polydimethylsiloxane (PDMS) microfluidics devices opening up a wider range of applications. A constant flow with a rate ranging from 0.8 nl/s to 1.8 nl/s was achieved by adjusting PDMS wall thickness.

Keywords: microfluidics, micropumping, vacuum-driven

1 INTRODUCTION

One of the widely studied sectors in microfluidic point-of-care testing is controllable, hand-held pumping [1-7]. In our previous study [1], syringe-assisted pumping utilizing the gas permeability properties of polydimethylsiloxane (PDMS) was introduced. The pumping can drive the liquid without external pumps, however, there were inherent drawbacks. First, the flow rate significantly decreased after the fluid entered the part of channel surrounded by the micro-chamber. This is because the air diffusion area decreased while the flow reached the dead-end channel surrounded by the vacuum-chamber, resulting in a non-constant flow rate. Second, previous pumps could not be used in the microfluidics system with integrated functions. This is because the vacuum chamber is pre-integrated in the device, leading to all of the channels being suspect to air diffusion, which can trigger samples into unwanted channels. In order to integrate multiple pumps, air diffusion from unwanted channel should be isolated. Third, a vacuum chamber was designed in-plane, which has limitation in terms of design flexibility. In this paper, we propose a compact syringe-assisted pumping method, which can achieve a constant flow rate. Multiple pumps can be integrated in one device, and no pre-determined vacuum chamber is needed.

2 THEORETICAL MODEL

One of the vacuum assisted pumping methodology is shown in Fig. 1 (a). The device consists of inlet, fluidic channel and a vacuum chamber which has a port to be connected with a syringe. Here, a dead-end channel is partly surrounded by an embedded microchamber, with a thin PDMS wall separating the dead-end channel and the embedded microchamber. The pumping works for simple designs, however, its embedded microchamber brings some limitations. First, the flow rate significantly decreased after the flow entered the part of channel surrounded by the micro-chamber. This is because the air diffusion area decreased while the flow reached the dead-end channel surrounded by a vacuum-chamber, resulting in a non-constant flow rate. Second, previous pumps cannot be used in the microfluidics system with integrated functions. Since the vacuum chamber is pre-integrated in the device, all of the channels will be suspect to air diffusion, which is not used with devices with integrated microfluidic components. In order to integrate multiple pumps, air diffusion from unwanted channel should be isolated. Third, a vacuum chamber was designed in-plane, which has limitation in terms of design flexibility.

Our proposed pumping is shown in Fig. 1 (b). The bottom layer consists of a traditional fluidic network, which has inlet, fluidic channel and outlet. The pump unit is bonded on the outlet. The pump consists of dead-end channel and vacuum chamber which has a syringe port. First, a syringe is connceted to the microchamber by a tube. By pulling a syringe plunger, a negative pressure will be generated inside the vacuum chamber, where cylindrical posts are placed to prevent the collapse of vacuum chambers. As PDMS is a gas-permable material, a pressure difference inside fluidic channel and vacuum chamber causes gas diffusion. Gas inside fluidic channel diffuses into vacuum chamber through thin PDMS wall. Then. liquid is loaded in the inlet after a few seconds to allow for a constant and steady air flux. As described in the paper [1], the flow rate Q can be rewritten by using gas-permeability of PDMS

$$Q(t) \approx \mathrm{k}\frac{FS}{\mathrm{C_{ATM}}} = \mathrm{kD}\frac{\mathrm{C_{PDMS}} - \mathrm{C_{Chamber}}}{\mathrm{C_{ATM}}}\frac{S}{t_{wall}}$$

$$\rightarrow Q(t) \propto \frac{S}{t_{wall}}$$

, where k is empirical factor related to viscous effect of the pumped liquid flow. F is steady state air flux diffusing into micro-chamber through PDMS wall. S is the total surface area that allows air to diffuse into the PDMS bulk, which is equivalent to overlap area. C_{PDMS}, $C_{Chamber}$ and C_{ATM} is air concentration in PDMS, micro-chamber and atmosphere respectively. t_{wall} is PDMS wall thickness, where we designed to be 100 μm, 150 μm and 200 μm. As described in the paper [1], in order to achieve constant and steady-state air flux across the PDMS wall, diffusion time across the PDMS wall is considered.

$$t_{D1} \approx w^2 D^{-1}$$

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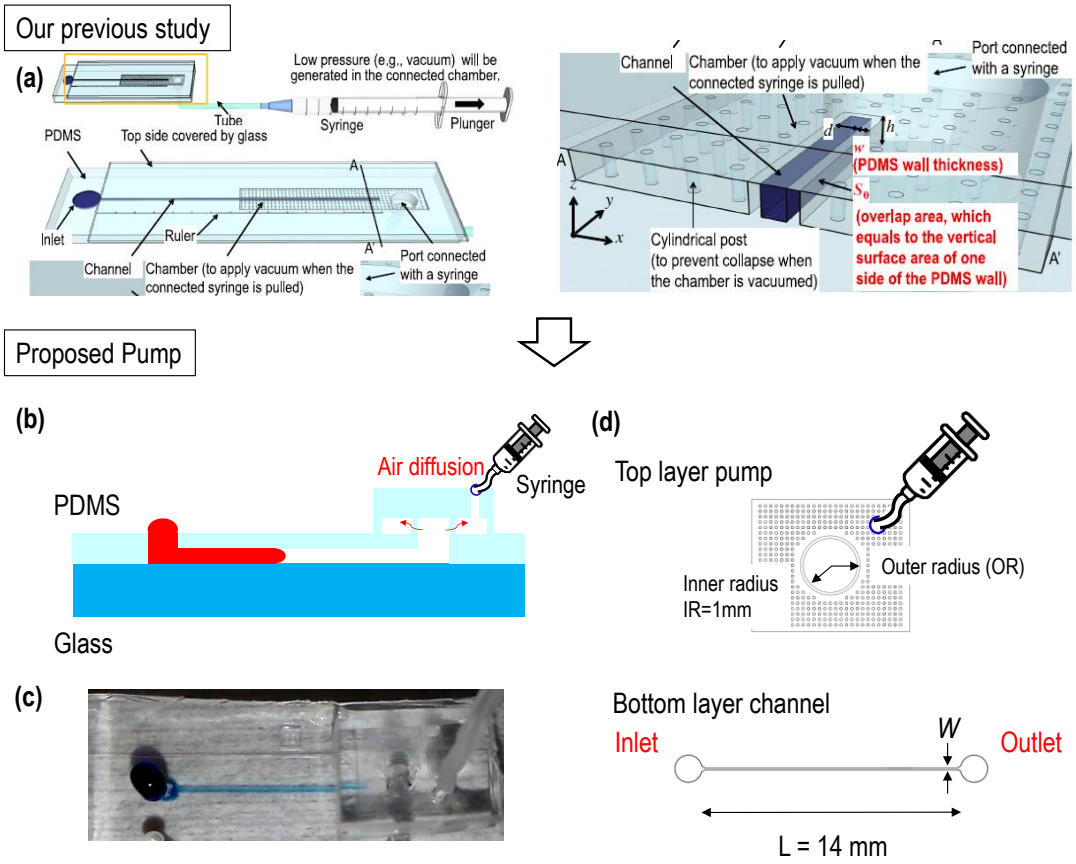

Figure 1: (a) Device schematic from our previous study [1], (b) schematic of proposed device (c) testing image, (d) design of the device. Bottom layer was designed to be a 14 mm length, 100 um width and 50 μm height straight channel. The outlet and inlet are designed to be 0.75 mm radius. Top layer is a proposed pump, which has inner radius with 1 mm radius and outer radius is varied by designing different wall thickness; 100 μm, 150 μm and 200 μm to study controlling flow rate.

If the PDMS wall thickness is w $= 100\mu m$, D $= 3.4 \times 10^{-9} m^2 s^{-1}$, the characteristic time to initialize the steady-state air flux is $t_{D1} \approx 2.94s$. Another characteristic time for diminishing air diffusion from the PDMS surface into the vacuum chamber across PDMS layer depends on its thickness. If PDMS thickness $w = 5mm$, the characteristic time is $t_{D2} \approx 2h$. In order to operate the device with constant and steady-state air flux, device needs to be operated in $t_{D1} < t < t_{D2}$.

Compared to the previous design, our proposed pump can be used with traditional microfluidics design. In addition, the pump itself can be designed independently from fluidic networks. On demand pumps can be chosen for fluidic networks. There was a limitation in the previous design that vacuum chamber was designed on the same layer with fluidic channel. For instance, the vacuum chamber has to be the same height as the microfluidics channel leading to difficulty in designng specific flow rates for pumping due to the fact that the geometry of vacuum chamber will determine the flow rate. Also, the vacuum chamber may be trigerring samples into unwanted channels in complex platform. By separate pumpings from a fluidic layer, microfluidic can be designed individually and pump can drag the samples in specific channel.

In this experimental sets, a bottom layer was designed to be a 14 mm length, 100 um width and 50 μm height straight channel. The outlet and inlet are designed to be 0.75 mm radius. Top layer is a proposed pump, which has inner radius with 1 mm radius and outer radius is varied by designing different wall thickness; 100 μm, 150 μm and 200 μm to study controlling flow rate.

3 DEVICE FABRICATION

The micropumps units and traditional fluidic channels were fabricated separately using photolithography and soft-lithography. In detail, first off, a Si mold for a fluidic channel was fabricated using traditional photolithography process. A 3-inch silicon wafer with one side polished (University wafers, South Boston, MA, USA) was submerged into buffered hydrofluoric acid (BHF) at room temperature for 5 min to remove the native silicon dioxide layer. Afterwards,

the 50µm thick master mold was fabricated with photolithography using SU-8 2050, a negative photoresist. In order to peel off the PDMS from the wafer mold easily, surface was treated with hexamethyldisilazane (Sigma Aldrich, Saint Louis, MO, USA). Next, uncured PDMS (mixing ratio of base polymer:curing agent = 10:1) was poured on the mold and cured at 100 ºC for 15 minutes following peeling off. Then the PDMS slab was bonded irreversibly on to glass substrates by oxygen plasma treatment using plasma cleaner PDC-32G (Harrick Plasma, Ithaca, NY, USA) at power of 18 W. Using the same procedure, micropump units can be fabricated with specified height. In our design, dead-end channel height was designed to be 50 µm. Lastly, micropump units were bonded on the outlet of fluidic channel. The final device picture is shown in Fig. 1 (c).

4 RESULTS AND DISCUSSION

Flow rates with three different PDMS wall thicknesses are shown in Fig. 2. Fig. 2 (a), (b) shows the device from previous study and its flow rates from the previous pumping, which has vacuum chamber on the same layer with fluidic channel. As pointed out in the theoretical model section above, the flow rate is propotional to the overlap area. Since the dead-end channel is designed in-plane with the fluidic channel, the overlap area decreases as the sample approaches to the dead-end channel. Therefore, the overall flow rate decreased. Fig. 2 (b) shows the flow rate from proposed pump. The flow rate is linear across the fluidic channel. Since the dead-end channel is designed as a part of the micropump, the overlap area is always constant while the sample flows in the fluidic channel. In other words, the sample is dragged toward the outlet with a constant force while flowing through the fluidic channel. Here, the thickness of the PDMS walls were designed to be 100 µm, 150 µm and 200 µm. Therefore, an overlap area can be calculated to be 0.66 mm^2, 0.67 mm^2 and 0.69 mm^2 for wall thickness at 100 µm, 150 µm and 200 µm respectively. Flow

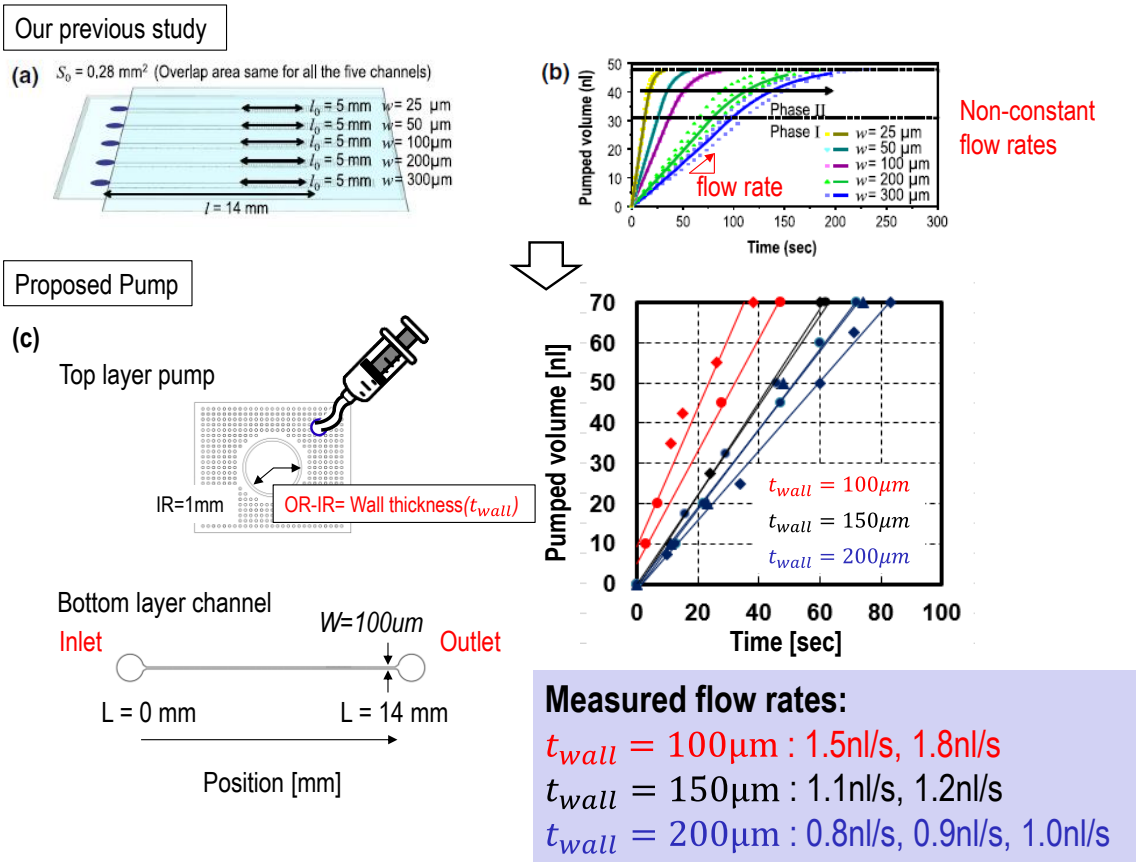

Figure 2: (a) Device schematic from our previous study [1] and (b) its flow rate, (c) proposed device schematic and (d) flow rate testing results. The PDMS wall thickness are designed to be 100 µm, 150 µm and 200 µm. Therefore, an overlap area can be calculated to be 0.66 mm^2, 0.67 mm^2 and 0.69 mm^2 for wall thickness at 100 µm, 150 µm and 200 µm respectively. Flow time is measured corresponding to the liquid position. As shown in the graph, the pump provided a constant flow rate despite liquid position. The flow rate is inverted proportional to the wall thickness.

time is measured corresponding to the liquid position. As shown in the graph, the pump provided a constant flow rate despite liquid position. The flow rate is inverted proportional to the wall thickness. resulting in a flow rate of 0.8 nl /s to 1.8 nl/s.

5 CONCLUSION

In conclusion, we proposed a syringe assisted, vacuum driven micropump providing a constant flow rate, which has the potential to be integrated into a variety of microfluidics system. The proposed pumping system does not require pre-designed vacuum chambers, allowing multiple pumps to be integrated into one device and enabling a constant flow rate. This method enables a point-of-care pumping system with more flexibility in hand-held PDMS microfluidics devices. A constant flow with a rate ranging from 0.8 nl/s to 1.8 nl/s was achieved by adjusting PDMS wall thickness.

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