Design and characterisation of a three-forked micropump on a fluid circulation channel

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A fluid circulation micropump that has three control channels activated by a single pressure source is reported. The proposed micropump can rotate a fluid in one direction because the control channels have widths that decrease in the fluid flow direction, thereby squeezing a fluid channel with different deflections under an identical pressure. The response time and deflection of the control channels are also investigated to demonstrate the effect of width differences on the flow rate of the micropump. In this reported work, several micropumps with different channel width ratios are fabricated, and their flow rates are experimentally analysed; the maximum flow rate of the proposed micropump was 10.8 nl/min at 3 Hz and 350 kPa for control channel widths of 300, 200 and 100 µm.

1. Introduction: One function of a lab-on-a-chip (LoC) is to automatically diagnose a variety of diseases using a single chip by conveying biomaterials to a specific target through microchannels. For biomaterial transport, LoCs often require a combination of micropumps and microvalves. Generally, micropumps can be classified into two categories: non-deflection types (electroosmotic, electrohydrodynamic, magnetohydrodynamic etc.) and deflection types (pneumatic, piezoelectric, electrostatic etc.) [1—3]. A microsystem that uses non-deflection micropumps can be constructed with a small size, can be simply integrated into LoC systems, and can have a long lifespan since they do not require moving parts for fluid transportation, although these micropumps may have problems associated with joule heating and bubble generation [2]. These two problems can degrade the properties of the biomaterials or interrupt the main fluid stream while the biomaterials move towards a target. Moreover, the performance of the non-deflection micropumps can be critically affected by the electrical properties of a fluid, such as the conductivity and permittivity. Therefore deionised (DI) water, which is a general fluid used to transport biomaterials, can be difficult to use in non-deflection micropumps since these two problems tend to occur with DI water. Meanwhile, deflection micropumps have been used for various bio applications even though their fabrication is not simple. The major advantages of these micropumps are that their actuation mechanisms are well defined and that they are not sensitive to the electrical properties of fluids. Thus, deflection micropumps can be easily designed, analysed and applied [3].

A typical deflection peristaltic micropump consists of three pneumatic control channels located on a fluid channel, in which the control channels are sequentially activated by either individual pressure sources or microvalves to induce the fluid flow [4]. It has been reported that the peristaltic micropumps have a relatively high pumping efficiency because of their lower leakage flow compared with other deflection micropumps such as the diffuser-nozzle and flap micropumps [5]. To date, peristaltic micropumps have used a variety of driving methods including piezoelectric, electrostatic and pneumatic sources.

Among these, the pneumatic pressure sources are commonly used to operate peristaltic micropumps for bio applications, because of negligible thermal damage to the sample fluid, provision of enough strength to squeeze the fluid channel and long lifespan compared with other driving methods [5]. The peristaltic micropumps activated by the pneumatic pressure sources are made by soft lithography that uses a polydimethylsiloxane (PDMS) material, which can provide flexibility, easy fabrication and biocompatibility [6, 7].

Several research groups have investigated peristaltic micropumps that operate with only a single pneumatic pressure source. For example, Wang and Lee [8] reported a new linear micropump with a serpentine-shaped channel instead of three control channels. The membranes located at the three intersections between the serpentine-shaped channel and the fluid channel was successively deflected resulting in a peristaltic flow when a pneumatic pressure was applied to the inlet of the serpentine-shaped channel. Lai and Folch [9] then demonstrated a linear micropump that was operated using a single-stroke (single pneumatic pressure source). They designed different widths for the control channels to deflect the fluid channel at different times. Recently, Lee and co-workers [10] conducted a feasibility study of a ‘three-forked’ circulation micropump that uses a single pneumatic pressure source.

In this Letter, we extensively characterise this micropump to verify the effectiveness of fluid circulation for use in a polymerase chain reaction (PCR). The principle of fluid circulation is investigated by measuring the deflection and response times of the membrane for various widths of the control channels. The flow rates are also analysed with regard to the operating frequency and the applied pneumatic pressure considering the diffusion effect.

2. Design of the three-forked micropump: The control channel width is a key design factor for regulating the response time of the membrane deflection, because a larger membrane is more flexible. Thus, a peristaltic effect can be generated in a fluid channel using several different control channel widths that are connected to the single pressure source. In addition, the membrane deflection is also dependent on the control channel width, as expressed in (1) [11]

$$\delta = \frac{q}{\pi^4 D\left(\left(1/a^2\right) + \left(1/b^2\right)\right)^2} \tag{1}$$

where δ is the deflection of the rectangular membrane, q is the

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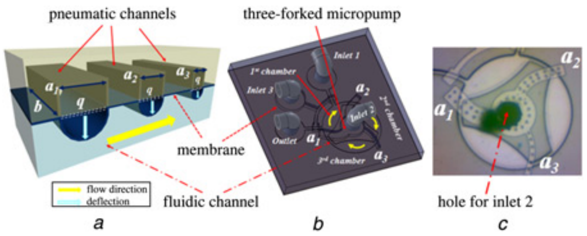

Figure 1 *Configuration of the three-forked micropump using a single pressure source*
a Principle of the three-forked micropump with the gradually increased/decreased widths of the control channels
b Proposed design of the three-forked micropump for fluid circulation
c CCD image of the proposed micropump

intensity of the uniform load, D is the flexural rigidity of the membrane, and a and b are the membrane width and length, respectively. The amount of deflection also affects the strength required to squeeze the fluid channel, which implies that the different control channel widths will induce fluid flow in one direction; the effects of the response time and deflection on the direction of the fluid flow are discussed in the Results and discussion Section. Fig. 1*a* presents a linear peristaltic pump with three membranes formed between each control channel and the fluid channel. When the control channels, whose widths differ from one another, are activated by a single pressure source, the micropump is expected to transport the fluid in the direction of the decrease in width.

The design and charged-coupled device (CCD) image of the proposed micropump for fluid circulation are shown in Figs. 1*b* and *c*. Three control channels of the micropump are located in a circle at 120° intervals. For a clockwise flow, the widths of the control channels are gradually decreased in the clockwise direction, with a certain width ratio (e.g. $a_1 = 300$ μm, $a_2 = 200$ μm and $a_3 = 100$ μm). The three control channels are connected using a single pneumatic pressure source through a microtube at the centre of the micropump. For PCR applications, the fluid circulation channel consists of three chambers and three interconnection channels. The total length of the fluid circulation channel is 940 μm at the centre, and the width of the three interconnection channels is 100 μm. A microvalve is also designed to control the on/off states of the inlet and outlet ports for fluid injection, and it is activated using a pneumatic pressure source that is applied through another microtube.

3. Fabrication and experimental setup: The proposed micropump was made from PDMS material, which provides superior flexibility and biocompatibility. Fig. 2 shows the fabrication sequence of the three-forked micropump using soft lithography. The fabrication process can be divided into three steps: mould preparation, replication and bonding.

For the mould preparation, a negative photoresist was spin-coated on a cleaned silicon substrate; then, it was patterned. The reflow process was performed at 180°C for 5 min to realise a semi-circular shape in the cross-section of the fluid channel, which enables the microvalve to effectively block the fluid flow in the off state [12].

For replication, the PDMS was mixed with a curing agent in a 10 : 1 ratio, and the bubbles were removed from the PDMS in a vacuum chamber. Through spin-coating at 2500 rpm and then curing at 85°C for 2 h in an oven, the first PDMS layer for the fluid channel and membrane was produced (Fig. 2*a*). Next, the mould of the second PDMS layer for the control channels (Fig. 2*b*) was positioned in a Petri dish, and the well-mixed PDMS was poured onto the mould. Finally, the curing process was conducted under the same conditions described in Fig. 2*a*.

For the bonding step, two PDMS layers were peeled from each mould, and the detached layers were bonded after an O_2 plasma treatment for 90 s. Then, the bonded PDMS layer was heat-treated at 110°C for 10 min to increase the bonding strength. All inlet and outlet ports for pneumatic or fluidic interconnection were punched at predetermined positions on the PDMS layers using a micro-puncher (inner diameter: 0.5 mm; Harris Uni-Core™, TedPella Inc., Sweden). The fabrication process of the proposed device was completed after bonding the PDMS layer and glass substrate using the 90 s O_2 plasma treatment, and then heating at 110°C for 10 min. The resulting thicknesses of all microchannels and membranes were 15 and 10 μm, respectively.

Microtubes (outside diameter: 0.61 mm; Intramedic Clay Adams, USA) were used to apply a pneumatic pressure to the control channels and to introduce a fluid into the fluid channel. To control the three-forked micropump and microvalve, the LabVIEW software (National Instruments, version 9.0, USA) was used. A vibrometer (MLD-211D, Nihon Kagaku Eng., Japan) and CCD camera (Inu-hightec, USB 2.0 CMOS, Korea) were used to measure the deflection and response time of the control channels and the flow rate of the three-forked micropump, respectively.

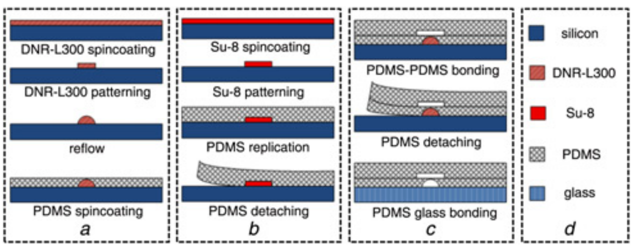

Figure 2 *Fabrication process of the three-forked micropump with soft lithography technology*

4. Results and discussion: To circulate the fluid, a three-forked micropump that uses a single pressure source was fabricated. First, the fluid channel was filled with DI water to evaluate the in situ characteristics of the fabricated micropump. The response time and deflection of the control channels were then measured using a vibrometer while a pneumatic pressure of 350 kPa was being applied to the three control channels for 0.3 s. Fig. 3 shows the response time and deflection measurements at the centre of the membrane with widths of 300, 200 and 100 μm. The falling time of the membrane was considered to be a meaningful response, because it was experimentally dominant compared with the rising time in a peristaltic micropump.

The response times of the 200 and 300 μm widths were similar (0.01 s), although the response time of the 100 μm width was notably slower (0.03 s). This result confirms that the response time of the control channel can vary considerably depending on the channel width within a certain range. In addition, the deflections

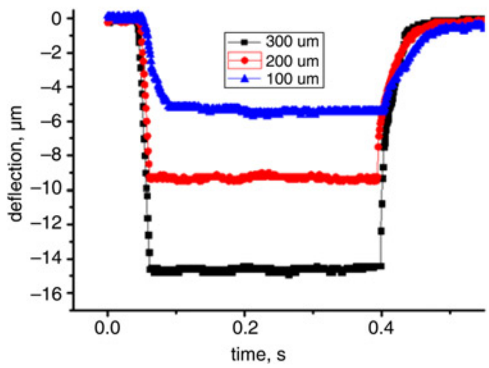

Figure 3 *Response time and deflection of membrane with respect to the widths of the control channels*

for the 300, 200 and 100 µm widths were measured to be 14.9, 9.5 and 5.75 µm, respectively. Note that in terms of the fluid flow, the flow direction was not affected by the difference in the response time alone because the peristaltic effect is not completely realised in the three-forked micropump. However, the amount of deflection becomes clearly different according to the membrane size, and proportionally generates the driving force for the fluid flow. Thus, a clockwise fluid flow can be achieved in the three-forked micropump by decreasing the membrane width in the clockwise direction.

An important performance capability in a micropump is the flow rate, which is defined as the volume of fluid that can be moved per unit of time. To evaluate the flow rate, two types of three-forked micropumps with different control channel width ratios were fabricated. The width ratio ($a_1 : a_2 : a_3$) of the type 1 micropump was 300 µm : 200 µm : 100 µm, and the ratio of the type 2 micropump was 300 µm : 150 µm : 50 µm. The experimental procedure is described as follows. The fluid channel was filled with DI water while the microvalve and micropump remained in an open state. To isolate the three chambers, the three-forked micropump was activated using a static pneumatic pressure of 350 kPa. Then, red ink (Pirot, UK) was injected to visualise the fluid flow until the DI water in the first chamber was completely replaced by the red ink. Next, the inlet and outlet ports of the fluid circulation channel were closed by the microvalve. The flow rate of the micropump was subsequently observed by the CCD camera while a pneumatic pressure ranging from 300 to 400 kPa was periodically applied to the micropump using the LabVIEW controls. This experiment was repeated with variations in the pumping frequency from 1 to 4 Hz.

To measure the flow rate, two images at specific time intervals were chosen from the successive photographic images. Image analysis using imaging software (Adobe Photoshop, version CS 3, USA) was used to obtain the flow rate considering the time interval of the two images, and the depth of the fluid channel.

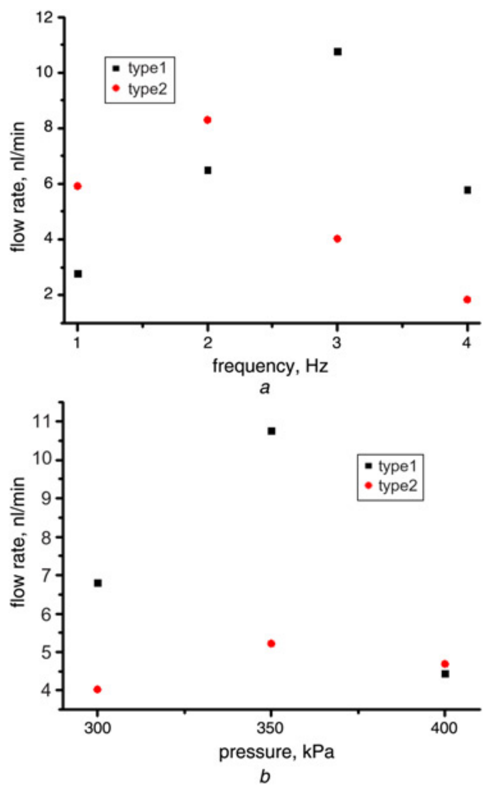

Figure 4 *Flow rates obtained from the micropump with different width ratios of the control channel*
a Flow rates with respect to the operating frequency of the micropump
b Flow rates with respect to the applied pressure to the micropump

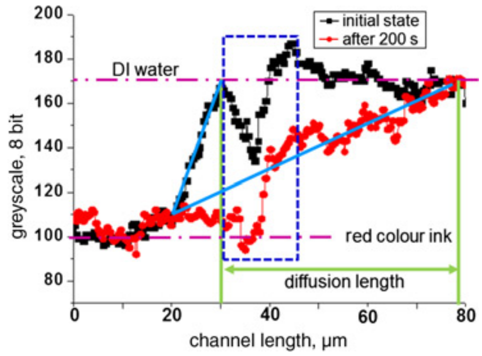

Figure 5 *Diffusion of the red ink to DI water at the interconnection channel between chambers*

Fig. 4*a* shows the measured flow rates, which decrease when the operating frequency exceeds 3 Hz. This can be theoretically inferred from the fact that the flow rate is expected to decrease as the operating frequency approaches 5 Hz, because the sum of the maximum falling and rising times of the membrane is about 0.2 s (Fig. 3). The response time becomes larger than expected because of the compliance of the fluid channels and the pneumatic microtube.

Fig. 4*b* shows the measured flow rates, which decrease when the pneumatic pressure exceeds 350 kPa. This can be explained by the fact that the membrane deflection towards the bottom of the fluid channel depends on the width of the control channel when the pneumatic pressure is <350 kPa. This operation corresponds to a width-controlled regime, in which the flow rate increases with the pneumatic pressure. However, when the pneumatic pressure is higher than 350 kPa, the membrane of the wide control channel is no longer deflected because the membrane already makes contact with the bottom of the fluid channel; whereas the membrane of the narrow control channel continues to go down depending on the pneumatic pressure, thus corresponding to a pressure-controlled regime.

The maximum flow rate of the type 1 micropump was 10.8 nl/min when the micropump was operated at 3 Hz with a pneumatic pressure of 350 kPa. In the type 2 micropump, the maximum flow rate achieved was 8.3 nl/min at 2 Hz and 350 kPa.

To investigate the diffusion effect on the flow rate of the three-forked micropump, the fluid channel was filled with DI water and red ink in the same manner as for the flow measurement. As soon as the pneumatic pressure applied to the micropump was released, the CCD images were taken for 200 s in the vicinity of the interconnection channel that includes a diffusion boundary line between the DI water and red ink. The diffusion status in an 8 bit grey scale is estimated for the two CCD images at 0 and 200 s, and this is displayed in Fig. 5, where the corresponding values for the red ink and DI water are indicated at 100 and 170 in the 8 bit grey scale, respectively. The estimation was conducted using the Visual C^{++} software (Microsoft, Visual Studio, version 2008, USA). The concentration of the ink in the first chamber appears to decrease according to the interconnection channel length, because the ink diffuses into DI water presenting in the interconnection channel with the passage of time. Note that the grey scale values in the dotted area were distorted because of shadow effect of the control channel. The starting point of the diffusion was selected when the first value was 170 in the first CCD image (0 s), and the end point was 170 in the second CCD image (200 s). The diffusion length, which is defined as the difference in the channel length between the starting and ending points, was 48.7 µm at 200 s; the diffusion length can be equivalently converted into a flow rate of 0.02 nl/min, which corresponds to only 1.9% of the maximum flow rate in the type 1 micropump.

5. Conclusion: A three-forked micropump operating with a single pneumatic pressure source is proposed for circulating fluid in a microchannel. The effect of the control channel width was confirmed by measuring the response time and deflection. To investigate the flow rate, several control channel width ratios were tested under various operating frequencies and pneumatic pressures. The maximum flow rate was calculated to be 10.8 nl/min at 3 Hz and 350 kPa. All experimental results support the finding that the proposed micropump can successfully circulate a fluid in one direction using a single pneumatic pressure. Therefore it is expected that the proposed micropump can be applied to PCR applications because of its simple operation and easy integration.

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7 References

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