Optimum design and investigation on diffuser polymethylmethacrylate (PMMA) peristaltic micropumps

Y. C. Hsu, N. B. Le, and M. S. Lin, and L. S. Jang

Abstract—Utilizing micro-electro-mechanical-systems (MEMS) techniques and a solvent-assisted bonding process, a new generation of diffuser peristaltic polymethylmethacrylate (PMMA) micropumps was optimized and fabricated. The main purpose of this study is to compare the performance of optimized and un-optimized micropumps which have the same diffuser throat/inlet area (i.e. 16000 μm²). Furthermore, an additional optimized design which has smaller diffuser inlet area was considered to validate and analyze the effect of diffuser inlet area to the micropump performance. The experimental results were validated by comparing with previous generation which had not been optimized the diffuser element. Specifically, the experimental results showed that, with similar diffuser element inlet area (i.e. 160000 μm²), with and without optimized micropumps yield maximum flow rates of 246.4 μL/min and 194.8 μL/min, respectively. Furthermore, it is shown that the back pressure in the optimized micropump is 6.9 kPa, while that in the un-optimized pump is 5.69 kPa. The effect of diffuser element throat/inlet area to pump flow rate and back pressure was investigated by comparing the experimental results of two optimized designs, one with 80 μm × 80 μm and the other with 127 μm × 127 μm cross-sectional area. The results indicated that, the design with larger inlet area gave higher flow rate. However, the rate of reduction in the maximum flow rate with increasing back-pressure increases at the higher inlet area design, which is due to the greater pressure dissipation/loss associated with a larger channel cross-sectional area.

I. INTRODUCTION

Peristaltic pumps have a number of crucial advantages for microfluidic applications. Firstly, they contain no mechanical moving parts, and thus the risk clogging and sample damage is substantially reduced. Moreover, the lack of mechanical moving parts increases both the reliability and the life expectanc of the micropump. In addition, peristaltic pumps have a planar design, which not only simplifies the fabrication process and reduces cost, but also enhances the flow rate, and thus improves the throughput of the microfluidic application. Finally, peristaltic pumps are suitable for the manipulation of both reagents and live cells, and are therefore a practical choice for a wide variety of microfluidic applications.

Compared to conventional materials (i.e. silicon, quartz or glass substrates), polymers offer a number of significant advantages, including low cost, chemical inertness, light weight, bio-compatibility, optical clarity, good mechanical strength, and so forth. Table 1 compares the principal advantages and disadvantages of silicon, glass and polymer as regards their suitability for microfluidic applications. Researchers have successfully demonstrated the use of a variety of polymers in the fabrication of microfluidic devices, most notably polydimethylsiloxane (PDMS) [1], polymethylmethacrylate (PMMA) [2] and polycarbonate (PC) [3]. Of these materials, PDMS is the most commonly used since it is readily processed using conventional MEMS-based techniques. However, PDMS has a low modulus of elasticity, and therefore the maximum driving force which can be developed by the pump is inevitably reduced. Furthermore, the relatively poor stiffness of PDMS reduces the back pressure within the micropump, and therefore limits its pumping power. For example, the PDMS peristaltic pump developed in [4] has a maximum back pressure of 16.5 mm-H₂O (0.14 kPa), which is considerably lower than that of the silicon-based pump presented in [5], i.e. 3.2 kPa. Compared to PDMS, PMMA has a greater stiffness, and can therefore increase the driving force within the pump. Nonetheless, PMMA has found only limited use in the fabrication of microfluidic devices in recent years due to the requirement for high bonding temperatures and pressures in the packaging process. However, in [6], the current group showed that these limitations could be overcome via the use of a solvent-assisted PMMA-to-PMMA chip-level bonding technique.

TABLE 1

| Qualitative comparison of silicon, glass and polymer in terms of their suitability for microchip applications |
|---|---|---|
| Feature | Silicon | Glass | Polymer |
| Cost | High | High | Low |
| Optical clarity | Bad | Good | Good |
| High-aspect-ratio structure | Slow | Difficult | Easy |
| Bio-compatibility | Bad | Bad | Good |
| Mechanical robustness | Bad | Bad | Good |

In conventional peristaltic micropumps, the actuation chambers are connected to one another (and to the inlet and...
outlets of the pump) using straight microchannels. While this configuration yields a high flow rate and a bi-directional pumping capability, it limits the back pressure generated within the pump, and therefore reduces the ability of the pump to drive viscous fluids such as human blood, insulin, and so forth. In an attempt to improve the back pressure performance of the conventional design, this study develops and optimizes a peristaltic micropump in which all of the straight microchannels are replaced by optimized diffuser elements. Using conventional MEMS-based techniques, two optimized designs of diffuser peristaltic micropump were fabricated on PMMA substrates and were then characterized experimentally in terms of their flow rate and back pressure properties.

II. DESIGN, FABRICATION AND OPERATING PRINCIPLE OF PERISTALTIC MICROPUMPS

A. Pumping operation

The diffuser peristaltic micropumps incorporate three serially-connected actuation chambers, each with a PZT-Glass actuator-membrane mechanism positioned above it. The application of an electrical voltage to the actuator causes the membrane to deflect in the upward direction, causing the pressure within the chamber to drop. When the voltage is removed, the membrane deflects in the downward direction, thereby restoring the pressure to its original value. Thus, by applying an appropriate voltage control scheme, the membranes can be actuated sequentially in such a way that fluid is drawn into the pump and transported peristaltically along its length to the outlet pipe.

Figure 1 illustrates the four-phase voltage control scheme applied to achieve the peristaltic pumping effect. In the first phase, a voltage is applied to the leftmost PZT actuator causing the glass membrane to deflect in the upward direction, thereby allowing fluid to flow into the first chamber. In the second phase, a voltage is applied to both the first and the second actuators causing both membranes to deflect in the upward direction and allowing the fluid to travel from the first chamber to the second. In the third phase, the voltage is removed from the first actuator and applied to both the second and the third actuators. Thus, the first membrane deflects in the downward direction, the second membrane remains in the upward direction, and the third membrane deflects in the upward direction. As a result, the fluid travels from the second to the third chamber. In the final phase, the voltage is removed from the second actuator and is applied only to the third actuator, causing the membrane to remain in the upward position. As a result, the fluid is expelled from the third chamber and exits the micropump via the outlet tube.

B. Design and fabrication

1) Optimum design for diffuser element

The diffuser element has a flow directing capability in the diffuser direction. Furthermore, diffuser element has the ability to reduce the velocity and increase the static pressure of fluid passing through a system [7]. In peristaltic micropump, the operation of actuator-membrane generates valving effect to direct the flow. However, due to the geometry of chamber, the valving efficiency generated by actuator-membrane is small. Consequently, directing capability of diffuser element is expected to increase the valving efficiency, thus, raising the flow rate. In addition, the ability to increase the static pressure in the reverse direction of diffuser element is predicted to increase the pressure difference (back-pressure) significantly. Therefore, optimum design process for obtaining high efficiency of diffuser elements is significant and essential. This section discusses on diffuser element optimization design process and from which define the optimum parameters for our pump design.

The diffuser element geometry and designed parameters are shown in figure 2. In figure 2 $W_1$ is throat/inlet width, $W_2$ is exit/outlet width, $L$ is element length, and $2\theta$ is divergence angle. In addition, $b$ (not shown) is element depth/height. The semicircles at both sides of the diffuser element are intentionally integrated because the depth of channels and diffuser valves compared to that of chambers are different. Thus, the semicircles help decrease the effect of the suddenly changing of cross-sectional area, which helps avoid the pressure loss and particle damage when operating.

The basic output of a diffuser is the pressure-recovery coefficient $C_p$, defined as [14]
\[ C_p = \frac{P_e - P_t}{P_{0e} - P_t} \]

Where, \( p \) is pressure and subscripts \( e \) and \( t \) denote the exit and the throat, respectively. The optimum design problem for diffuser element is to maximize the pressure-recovery coefficient and this coefficient depends upon diffuser element’s geometry through the following parameters: divergence angle \((2\theta)\), slenderness \((L/W_1)\), and aspect ratio \((AS=b/W_1)\).

Many researches [9]–[11] were reported experimentally and theoretically on the optimization of the divergence angle \((2\theta)\) by drawing the relation between diffuser element efficiency and divergence angle. Those researches suggested the divergence angle varies from 5° to 10°. Therefore, repeated work is not necessary and designer can choose a value in that range. In this study, a divergence angle of 7° was chosen. The second parameter need to be designated is inlet area \((A_1)\). Inlet area is designated with a carefully considering of compression ratio, chamber geometry, system flow rate, and back-pressure. Specifically, for a constant excitation frequency, the flow rate increases with an increasing inlet area provided that its value does not exceed the cross-sectional area of the chambers and the inertia effects are properly controlled. Furthermore, for a constant backpressure, the flow rate increases with an increasing inlet area. However, the flow rate reduces more rapidly with an increasing back pressure at higher values of the inlet area. This assumption was validated by experimental results in section 3. In this study, the inlet area of 80 µm × 80 µm and 127 µm × 127 µm were chosen. These values were assigned with the aspect ratio of 1 which was investigated, optimized and suggested in references [8], and [11]. Consequently, the inlet widths of the two designs are 80 µm and 127 µm, respectively.

After choosing the divergence angle, the slenderness \((L/W_1)\) was defined by using the map in figure 3 with the purpose of maximizing the pressure-recovery coefficient \((C_p)\). The diffuser element length \((L)\) and outlet width were obtained by utilizing the slenderness value, inlet width, and the diffuser element geometry. Table 2 tabulates two set of optimized parameters of our diffuser element. These parameters were then utilized to design the diffuser elements shown in figure 4.

<table>
<thead>
<tr>
<th>Design #</th>
<th>2θ (degree)</th>
<th>( W_1 ) (µm)</th>
<th>( b ) (µm) (controlled by etching time)</th>
<th>( \frac{b}{W_1} )</th>
<th>( L ) (µm)</th>
<th>( L/W_1 )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>7</td>
<td>80</td>
<td>80</td>
<td>1</td>
<td>1520</td>
<td>19</td>
</tr>
<tr>
<td>2</td>
<td>7</td>
<td>127</td>
<td>127</td>
<td>1</td>
<td>2413</td>
<td>19</td>
</tr>
</tbody>
</table>

![Figure 4 Diffuser element optimum designs, a) 80 µm inlet width, and b) 127 µm inlet width.](image)

2) Fabrication

![Figure 5 Schematic overview of PMMA peristaltic micropump fabrication process.](image)

Figure 5 presents a schematic illustration showing the basic steps involved in the fabrication diffuser-type micropumps. The fabrication process commenced by using a sputtering process to deposit a thin (2000 Å) aluminum adhesive layer on the upper surfaces of two silicon wafers,
each with a thickness of 500 μm. Using a conventional photolithography technique and a deep reactive ion etching (DRIE) process, the two substrates were then patterned with the required microchannel and chamber configurations of the two micropumps (figure 5(a)). The two designs of diffuser elements have a throat/inlet width of 80 μm and 127 μm and a length of 1520 μm and 2413 μm, respectively (see figure 4 and figure 6). Having patterned the microchannels on the two wafers, the process was repeated to pattern each wafer with three uniformly-spaced circular chambers with a depth of 15 μm and a diameter of 7 mm. A thin gold layer was then evaporated onto each patterned silicon wafer (figure 5(b)). Using the patterned silicon wafer as a mold, a complementary nickel wafer was fabricated using an electroforming system (figure 5(c)). The patterned nickel wafer (figure 5(d)) was then used to replicate the structure of the original silicon wafer in a PMMA wafer by performing a hot embossing process at a temperature of around 130 °C (figure 5(e)). Having done so, three square openings were carved into a blank PMMA wafer using a CO2 laser beam to accommodate the actuator-membrane mechanisms (discussed below). Finally, the upper and lower PMMA wafers were aligned (figure 5(f)) and bonded using the solvent-assisted PMMA-to-PMMA technique described in [6] (figure 5(g)). Figure 7 presents a photograph of the bonded PMMA micropump structure. In the final stage of the assembly process, PZT actuators (Piezo Systems, Inc., T107-H4E-602) with a length and width of 7 mm and a thickness of 235 μm were attached to three glass membranes with a thickness of 150 μm using epoxy glue. The actuator-membrane structures were then integrated with the pump body. Figure 8 presents a photograph of the completed peristaltic micropump.

III. EXPERIMENTS AND RESULTS

A. Experimental setup

Figure 9 Schematic illustration of experimental setup.

Figure 9 presents a schematic illustration of the experimental setup used to characterize the flow rate and back pressure properties of the two micropumps. During the experimental trials, the pumps were actuated at various frequencies and driving voltages using a voltage control system driven by a function generator and a differential amplifier. The resulting membrane displacement was measured using a fiber-optic displacement system (MTI Instruments, MTI 2000 Fotonic Sensor). The signals from the four-phase controller, the differential amplifier and the fotonic sensor, respectively, were recorded and displayed using a digital oscilloscope. The pumping tests were all performed using DI water as a working fluid. During the pumping operation, the fluid oscillates back and forth within the micropump as a result of the pulse-like actuation effect. Thus, the flow rate of the pump actually represents the net movement of the fluid in the forward direction. In the current experiments, the pumping rate was estimated by using a microbalance to measure the change in weight of the pumped fluid over a specified elapsed time.

The experimental trials commenced by measuring the displacement of the PZT-glass actuator-membrane as a function of the driving frequency. As shown in figure 9, the fotonic sensor was positioned such that it was aligned immediately above the center point of the middle actuator-membrane. In the tests, the peak-to-peak displacement of the membrane was measured at a constant driving voltage of 100 Vpp and frequencies ranging from 10 Hz to 1 kHz. The corresponding results are presented in figure 10. As shown, the membranes exhibit a maximum peak-to-peak displacement of 0.81 μm at a driving frequency of 150 Hz.
B. Flow rate and back pressure characteristics of optimized and un-optimized diffuser peristaltic micropumps

Figure 11 illustrates the variation in the flow rates of the three micropump designs as the driving frequency is increased from 10 Hz to 1 kHz at a constant driving voltage of 100 V<sub>pp</sub>. As shown, the second optimized design micropump yields a significantly higher flow rate than the first optimized design. Specifically, the first design with 80 μm × 80 μm diffuser element inlet area has the maximum flow rate of 128.4 μL/min at 100 Hz, while the second design with 127 μm × 127 μm diffuser element inlet area yields 246.4 μL/min maximum flow rate at 150 Hz. The lower flow rates of the first design could result from the smaller geometry of the channels. Usually, smaller channel sizes would result in lower flow rates. Figure 11 also demonstrates the comparison between the optimized and un-optimized designs which have similar diffuser inlet area (i.e. 16000 μm<sup>2</sup>) and the results show a higher performance of the optimized design. That is, the optimized design has the maximum flow rate of 246.4 μL/min, while the un-optimized design has the maximum flow rate of 194.8 μL/min.

The back pressure developed by a micropump provides an indication of its pumping power. In general, a higher back pressure is beneficial since it enables the pump to transport fluids with a greater viscosity. However, the small characteristic size of the microchannels and actuation components used in micropumps restricts the maximum value of the back pressure which can be attained. In practice, the maximum back pressure of a pump can be measured by gradually raising the height of the output reservoir relative to the pump (see figure 12) and measuring the height at which the flow rate reduces to zero. Figure 13 illustrates the correlation between the flow rate and the back pressure in the current micropumps when driven by a constant driving voltage of 100 V<sub>pp</sub> and an excitation frequency of 150 Hz. From inspection, it is determined that the maximum back pressures of the first design (i.e. 80 μm × 80 μm) inlet area is 1134 mmH<sub>2</sub>O (11.12 kPa), while that of the second design (i.e. 127 μm × 127 μm) is 707 mmH<sub>2</sub>O (6.94 kPa). The maximum back-pressure of the un-optimized design is found to be 580 mmH<sub>2</sub>O (5.69 kPa).

Figure 12 Experimental setup used to measure back pressure of micropumps.

Figure 13 Variation of maximum flow rate with back pressure for driving voltage of 100 V<sub>pp</sub> and excitation frequency of 150 Hz.
Table 3 summarizes the experimental results obtained for the flow rate and maximum back pressure characteristics of three micropump designs. Overall, the results show that the second optimized design micropump (i.e. 127 µm x 127 µm) provides a higher flow rate but has a lower maximum back pressure compared to the first design which has smaller inlet area (i.e. 80 µm x 80 µm). The higher flow rate is the result of larger inlet area which reduces the channel resistance. However, the larger channel dimension generates larger pressure dissipation/loss, therefore, yields faster decreasing slope of back-pressure versus maximum flow rate curve which result a smaller maximum back-pressure. Furthermore, the comparison of the optimized and un-optimized designs which has similar inlet area demonstrates significantly improvements in both flow rate and maximum back-pressure. The higher performance should be the result of higher diffuser element’s efficiency which was obtained from optimization process.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Optimized 1st design</th>
<th>Optimized 2nd design</th>
<th>Un-optimized design</th>
</tr>
</thead>
<tbody>
<tr>
<td>L</td>
<td>1520 µm</td>
<td>2413 µm</td>
<td>200 µm</td>
</tr>
<tr>
<td>Applied voltage</td>
<td>100 V&lt;sub&gt;PP&lt;/sub&gt;</td>
<td>100 V&lt;sub&gt;PP&lt;/sub&gt;</td>
<td>100 V&lt;sub&gt;PP&lt;/sub&gt;</td>
</tr>
<tr>
<td>Membrane max. displacement</td>
<td>0.81 µm (150 Hz)</td>
<td>0.81 µm (150 Hz)</td>
<td>0.81 µm (150 Hz)</td>
</tr>
<tr>
<td>Max. flow rate (µL/min)</td>
<td>128.4 (100 Hz)</td>
<td>246.4 (150 Hz)</td>
<td>194.8 (400 Hz)</td>
</tr>
<tr>
<td>Max. back-pressure (mmH&lt;sub&gt;O&lt;/sub&gt;)</td>
<td>1134 (11.12 kPa)</td>
<td>707 (6.94 kPa)</td>
<td>580 (5.69 kPa)</td>
</tr>
</tbody>
</table>

IV. CONCLUSION

This study has successfully demonstrated the fabrication of peristaltic micropumps on PMMA substrates. In the fabrication procedure, the microchannels and chambers are patterned using conventional MEMS-based techniques, including photolithography, DRIE etching, and hot embossing. The patterned substrate is then fused to an upper cover plate using a low-temperature, low-pressure, solvent-embossing. The patterned substrate is then fused to an upper cover plate using a low-temperature, low-pressure, solvent-embossing. Furthermore, the comparison of the optimized and un-optimized designs which has similar inlet area demonstrates significantly improvements in both flow rate and maximum back-pressure. The higher performance should be the result of higher diffuser element’s efficiency which was obtained from optimization process.

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REFERENCE