Transport of fluid
by magnetically actuated micropump
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by magnetically actuated micropump

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Approved by

[Signature]
Professor Do Hyun Kim
자기력으로 구동되는 미세유체펌프에 의한 유체의 전달

김 정 애

위 논문은 한국과학기술원 석사학위논문으로 학위논문심사위원회에서 심사 통과하였음.

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Abstract

A magnetically actuated micropump for controlling liquid samples in microfluidic devices is presented. The pump is fabricated using soft lithographic technique with polydimethylsiloxane (PDMS). The micropump is composed of diffuser, nozzle and micro chamber. The diaphragm of microchamber includes a NdFeB magnet, and the pump is actuated by a permanent magnet of a magnetic stirrer.

With respect to design of micropump, pumps with various dimensions (neck width, length, and divergence angle of diffuser element) were tested. And effect of elastomeric property of diaphragms was investigated by differing PDMS mixing ratios.

The magnetically actuated micropump induces reciprocating fluid flow. Frequency and back pressure dependency of pumping performance were examined. And various fluids were utilized as working fluids to investigate the relation between pump performance and viscosity. Unlike the other actuating methods, magnetic actuation by permanent magnet makes the diaphragm distorted. Therefore the pump rate changes depending on rotating direction of the external magnet. Finally, durability of PDMS diaphragm was tested. The initial pump performance lasted for enough time for microfluidic application, although swelling of PDMS occurred slowly.
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1. Introduction

Microfluidics is the research area manipulating fluids with length scales less than a millimeter. It is now widely recognized to have high potential towards various applications. The number of papers in journals concerning the microfluidic applications has been increasing rapidly since the last decade [Kamholz, 2004; Stone et al., 2003]. Various functional elements including valves, pumps, actuators, switches, sensors, dispensers, mixers, filters, separators, heaters, etc. are reported and they are combined to achieve a microfluidic network system which is called µTAS (micro Total Analysis System) or ‘Lab-on-a-chip.’

Micropump, which is used for fluid delivery application, is one of the essential devices for integrated micro systems. Various studies have been carried out about effective design of micropumps, analysis of microfluidic motion, and microfabrication. Among the various micropumping technologies, mechanical micropumps with vibrating diaphragms have generated the most interest although many novel pumping strategies such as pumps based on electrohydrodynamics or electroosmosis have also been presented.

Early efforts in fabrication of diaphragm displacement micropumps utilized check valves. These valves suffer from many problems such as high-pressure drops, wear, and fatigue with long-term usage. To overcome these problems, valveless diffuser-and nozzle-based micropumps have been introduced. These pumps utilize the different characteristics of pressure drop through a nozzle and a diffuser to direct the flow in one preferential direction, and hence cause a net pumping action.

Actuation of the membrane can be carried out in several ways; pneumatic,
thermopneumatic, piezoelectric, electrostatic, bimetallic, shape-memory alloy, and so forth. The common drawback of those actuating methods is found in complexity of microstructure. Therefore a novel actuating method using a permanent magnet is introduced in this research. A magnet in the diaphragm responds external magnetic field. And the diaphragm vibrates correspondingly to the change of magnetic field. There is no need of physical connection between micropump and external facilities for providing actuation power. In this research, simple fabrication method of a valveless micropump using poly dimethylsiloxane was investigated. And the fabricated micropump was actuated using a permanent magnet. Characteristics of the magnetically actuated micropump were studied.
2. Theoretical background

2.1. Principle of actuation

2.1.1. Various actuating methods

Pumps are generally categorized into two groups as Figure 2.1 shows: (1) displacement pumps, which exert pressure forces on the working fluid through one or more moving boundaries and (2) dynamic pumps, which continuously add energy to the working fluid in a manner that increases either its momentum (as in the case of centrifugal pumps) or its pressure directly (as in the case of electroosmotic and electrohydrodynamic pumps). Many displacement pumps operate in periodic manner, incorporating some means of rectifying periodic fluid motion to produce net flow. Such periodic displacement pumps can be further broken down into pumps that are based on reciprocating motion, as of a piston or a diaphragm, and pumps that are based on rotary elements such as gears or vanes.

The majority of reported micropumps are reciprocating displacement pumps in which the moving surface is a diaphragm. These are sometimes called membrane-actuated pumps or diaphragm pumps.

Reciprocating displacement micropumps with a wide range of designs have been reported. Actuator types and configurations vary widely; reciprocating displacement micropumps with piezoelectric, electrostatic, thermopneumatic, pneumatic actuations, etc., have been reported.

Figure 2.2 (a) and (b) illustrate piezoelectric actuators in lateral and axial configurations. The free strain that can be produced in the actuator places an upper limit on the stroke volume of a piezoelectric-driven micropump. The available driving voltage and the polarization limit of the piezoelectric material, in turn, determine the maximum piezoelectric free strain. In Figure 2.2 (b), the axial strain induced in the disk by applying an external axial
Figure 2.1  Classification of micropumps [Laser and Santiago, 2004].
Figure 2.2 Reciprocating displacement micropumps with various actuators; (a) piezoelectric actuator in the lateral-strain configuration, (b) piezoelectric actuator in the axial-strain configuration, (c) thermopneumatic actuator, (d) electrostatic actuators and (e) external pneumatic actuator [Laser and Santiago, 2004].
electric field causes the pump diaphragm to deflect, expanding and contracting the pump chamber.

Thermopneumatically actuated micropumps have another chamber above the primary pump chamber which holds a secondary working fluid in Figure 2.2 (c). The temporal response of thermopneumatic actuators is limited by the rate of heat transfer into and out of the secondary working fluid, and so thermopneumatically actuated reciprocating displacement micropumps typically operate at relatively low frequencies.

Electrostatic forces are widely used for actuation in MEMS devices [Teymoori et al., 2005]. An electric field between the membrane and a fixed counter electrode attracts the membrane and thus deflects it.

Reciprocating displacement micropumps actuated pneumatically, as shown in Figure 2.2 (e), have been reported. These pumps require an external pneumatic supply and one or more high-speed valve connections. Therefore they are not comparable to micropumps with fully integrated actuators.

2.1.2. Actuation using magnetic force

Electromagnetically actuated micropumps are often reported. Jackson et al.[2001] fabricated the iron powder-mixed PDMS membrane and actuated it electromagnetically for application of microvalves and micropumps. And Khoo and Liu [2000] electroplated Permalloy on PDMS membrane and actuated electromagnetically. Also reported is a electromagnetically rotating motor for micromixing [Barbic et al., 2001].

In this research, the diaphragm of micropump includes a small magnet on which magnetic poles are orientated vertically. A magnet stirrer commonly used for mixing solution was used to provide magnetic force periodically. The stirrer rotates a donut-like permanent
magnet on which two magnetic poles place half and half as Figure 2.3 shows. When attractive (or repulsive) force is induced between the magnet of micropump and the one of stirrer, the diaphragm of micropump gets bent downward (or upward). Thus, supply and pump modes are observed periodically. Frequency of actuation is controllable in the range of 60~1100 rpm using the dial of the stirrer.

The main advantage of this micropump is structural simplicity. Electrodes are unnecessary, therefore fabrication becomes simple and a disposable microsystem is possible to realize.

Figure 2.3 Schematic diagram of micropump actuation using a magnetic stirrer: Top view (upper) and side view (under).
2.2. Valveless micropump with diffuser and nozzle

The valveless pump is a diaphragm pump and the passive check valves are replaced by diffuser and nozzle as flow-rectifying elements. The valveless diffuser pump consists of two diffuser elements connected to a pump chamber with an oscillating diaphragm. In general, the diffuser/nozzle acts to decelerate or accelerate fluid flow as fluid traverses along its changing cross-sectional area.

A diffuser is an expanding duct and a nozzle is a converging duct. The flow-directing action in the pump is based on the fact that a diffuser/nozzle element can be geometrically designed to have a lower pressure loss in the diffuser direction than in the nozzle direction, for the same flow velocity.

During increasing the chamber volume (supply mode), the inlet element acts as a diffuser with a lower flow restriction than the outlet element, which acts as a nozzle. This means that a larger volume is transported through the inlet into the chamber than through the output, as shown in Figure 2.4 (a).

During decreasing chamber volume (pump mode) the outlet element acts as a diffuser with a lower flow restriction than the inlet element, which act as a nozzle. This means that a larger volume is transported though the outlet of the chamber than through the inlet, as Figure 2.4 (b).

The result for a complete pump cycle is therefore that a net volume is transported from the inlet side to the outlet side.

The main advantages of the diffuser pump are the absence of moving valves. Wear and fatigue in valves are eliminated. And the absence of moving structures in fixed-geometry valves may be advantageous when the working fluid contains cells or other materials prone to damage or clogging.
Figure 2.4  Operation of diffuser and nozzle-based micropump: (a) supply mode; (b) pump mode.
2.2.1. Theory for pumped volume

The pressure drops across a diffuser and a nozzle can be written as

$$\Delta P_d = \frac{\rho v_d^2}{2} \xi_d$$  \hspace{1cm} (1)$$

and

$$\Delta P_n = \frac{\rho v_n^2}{2} \xi_n$$  \hspace{1cm} (2)$$

where \( \rho \) is the fluid density and \( v_d \) and \( v_n \) are the fluid flow velocities in the narrowest part (the neck) of the diffuser and the nozzle. \( \xi_d \) and \( \xi_n \) are pressure loss coefficients of the diffuser and the nozzle, respectively.

Stemme and Stemme [1993] derived the equation for relation of stroke volume and pumped volume per one cycle as followed.

$$V_p = 2\Delta V \left[ \frac{(\eta_{nd})^{1/2} - 1}{(\eta_{nd})^{1/2} + 1} \right]$$  \hspace{1cm} (3)$$

where \( V_p \) is net pumped volume, \( \Delta V \) is the volume variation amplitude and \( \eta_{nd} = \xi_n / \xi_d \)

Here the bracketed part of the upper equation can be defined as the static rectification efficiency, \( \chi \) :

$$\chi = \frac{(\eta_{nd})^{1/2} - 1}{(\eta_{nd})^{1/2} + 1}$$  \hspace{1cm} (4)$$
2.2.2. Pressure loss coefficients

The diffuser elements can be divided into three regions, entrance, diffuser, and exit like Figure 2.5. In entrance and exit, sudden contraction and expansion occur contrary to gradual expansion in diffuser region. Thus the pressure drop can be expressed as bellow;

\[ \Delta P_{d,t} = \Delta P_{d,en} + \Delta P_d + \Delta P_{d,ex} \]  \hspace{1cm} (5)

The total pressure loss coefficient for the diffuser can be calculated as

\[ \xi_d = \frac{\Delta P_{d,t}}{\rho v_1^2 / 2} = \frac{\Delta P_{d,en}}{\rho v_1^2 / 2} + \frac{\Delta P_d}{\rho v_1^2 / 2} + \frac{\Delta P_{d,ex}}{\rho v_1^2 / 2} \]

\[ = K_{d,en} + K_d + K_{d,ex} \frac{A_1^2}{A_2^2} \] \hspace{1cm} (6)

Similarly, the total pressure loss coefficient for the nozzle is

\[ \xi_n = (K_{n,en} + K_n) \frac{A_1^2}{A_2^2} + K_{n,ex} \]

\hspace{1cm} (7)

Therefore, diffuser-nozzle efficiency can be written as

\[ \eta_{nd} = \frac{\xi_n}{\xi_d} = \frac{(K_{n,en} + K_n)(A_1^2 / A_2^2) + K_{n,ex}}{K_{d,en} + K_d + K_{d,ex}(A_1^2 / A_2^2)} \] \hspace{1cm} (8)

For pressure loss coefficients of sudden expansion and contraction, empirical data are available [White, 1999]. And pressure drop coefficients for gradual expansion and contraction in diffuser and nozzle are theoretically estimated by using lubrication approximation. If neck width of diffuser \( W_1 \) is slightly smaller than width of widest part of
diffuser $W_2$ in Figure 2.6, pressure of the cross-sectional area of diffuser can be supposed as a function of $x$ and velocity as a function of $y$. And velocities toward $y$- and $z$-directions can be supposed to be zero.

Since gravity is negligible in micro-scale, pressure-velocity relation is described as following.

\[
-\frac{dP}{dx} + \mu \frac{d^2 u}{dy^2} = 0 
\]  \hspace{1cm} (9)

B.C.

\[
\frac{du}{dy} = 0 \quad \text{at} \quad y = 0 \\
u = 0 \quad \text{at} \quad y = \pm W / 2
\]

$W$ is width of diffuser as $W = W(x) = W_1 + 2x \tan(\alpha / 2)$.

Then, velocity is obtained as

\[
u(y) = \frac{1}{2\mu} \left( \frac{dP}{dx} \right) \left( y^2 - \frac{W^2}{4} \right)
\]  \hspace{1cm} (10)

where $\mu$ is viscosity of working fluid.

When volumetric flow rate $\Phi$ is evaluated by integration of $u(y)$, it becomes

\[
\Phi = \int u(y) \, dA \\
= \int_{-W/2}^{W/2} \frac{1}{2\mu} \left( \frac{dP}{dx} \right) \left( y^2 - \frac{W^2}{4} \right) h \, dy 
\]  \hspace{1cm} (11)

\[
= -\frac{h W^3}{12\mu} \left( \frac{dP}{dx} \right)
\]

and $h$ is height of pump chamber.
\( \Phi \) is determined by stroke of diaphragm and is independent to the geometry. Pressure drop of diffuser direction is obtained as follows.

\[
\int_{b}^{a} dP = \int_{b}^{a} -\frac{12 \mu \Phi}{h} \left( W_1 + 2x \tan \frac{\alpha}{2} \right)^{-3} dx
\]  

(12)

\[
P_2 - P_1 = \frac{12 \mu \Phi \zeta (1 + \zeta \tan(\alpha/2))}{h W_1^2 (1 + 2\zeta \tan(\alpha/2))}
\]  

(13)

\( \zeta \) is dimensionless variable of \( L/W_1 \).

Then, pressure drop coefficient of diffuser direction \( K_d \) is

\[
K_d = \frac{P_1 - P_2}{\frac{1}{2} \rho \bar{u}_1^2}
\]

= \frac{24 \zeta (1 + \zeta \tan(\alpha/2))}{Re (1 + 2\zeta \tan(\alpha/2))}

(14)

where \( Re \) is Reynolds number.

Similarly, pressure drop coefficient of nozzle direction \( K_n \) is evaluated as

\[
K_n = \frac{P_2 - P_1}{\frac{1}{2} \rho \bar{u}_2^2}
\]

= \frac{24 \zeta (1 + \zeta \tan(\alpha/2))}{Re}

(15)
Figure 2.5  Definition of diffuser and nozzle directions.
Figure 2.6 Top-view of a diffuser geometry, where $L$ is the diffuser length, $\alpha$ the divergence angle, and $W_1$ the neck width. $P_1$ and $P_2$ indicate pressure at narrowest and widest parts respectively.
2.2.3. Compression ratio

The working principle of diaphragm pumps can be described by a cyclic process consisting of alternate supplying (the diaphragm stroke enlarges the pump chamber volume) and pumping (the diaphragm stroke reduces the pump chamber volume) phases, respectively. The ratio between the stroke volume $\Delta V$ and the dead volume $V_0$ defines the compression ratio $\varepsilon$:

$$\varepsilon = \frac{\Delta V}{V_0}$$

(16)

Because of the small stroke of micro actuators and the unavoidable large dead volume, the compression ratio is usually small. The electrostatically actuated micropump and two kinds of the piezoelectrically actuated micropumps which were reported by Richter et al. [1998] showed the compression ratio of 0.002, 0.017 and 0.085, respectively.
2.3. PDMS-based micropump

The most common method for fabricating micropumps is micromachining of silicon bonded to glass layers. However a number of reciprocating displacement micropumps has been fabricated by other methods than traditional silicon/glass micromachining. Improvements in techniques for fabricating precision components from polymers have led to increased use of polymers in reciprocating displacement micropumps. One of the most promising approaches to develop microfluidic components is the use of printing and molding techniques applied to soft polymeric materials, particularly poly (dimethylsiloxane) (PDMS). Characteristics such as bio- and chemical compatibility, mechanical robustness, and cheap fabrication process have made it popular for use in microfluidic devices [McDonald et al., 2000]. The chief advantage of this approach is the potentially low fabrication cost, which is an important factor for widespread adoption of any technology. Since the pump diaphragm comes into contact with the working fluid, however, the stability of soft polymer diaphragms such as swelling phenomena is a concerning issue [Lee et al., 2003].
3. Experimental methods

3.1. Fabrication of diffuser/nozzle-based micropump

Procedure to fabricate a micropump using PDMS is described as follows. First, the pattern is transferred onto Si wafer using typical photolithographic method. The negative photoresist, SU-8 50 (Microchem corp.) was spin-coated on a 4 inch silicon wafer in two steps. In the first step, SU-8 was poured on a wafer and the wafer was spun for 30 seconds at 500 rpm to spread SU-8 over the wafer because SU-8 is highly viscous. Then the wafer was spun for 30 seconds at 1500 rpm so that the thickness of the photoresist is uniform. Soft baking process followed the coating to remove solvent from the photoresist and harden the remaining photoresist. The wafer was heated on the hot plate at 90°C for 7 minutes. After being cooled to room temperature, the photoresist was covered with the mask which was printed on a transparent film and exposed to UV light. Exposure time was 60 seconds. Figure 3.1 shows the negative mask. Post-exposure baking was done at 90°C for 5 minutes. Then the photoresist was developed using SU-8 developer solution and washed with isopropyl alcohol.

The fabrication steps from now on are illustrated in Figure 3.2. After the expected pattern was made on the wafer through the procedure described above, PDMS (Dow Corning SYLGARD® 184) was prepared as mixed solution of base and curing agent in a ratio of 10 to 1. Before pouring it on the wafer, a cylindrical supporting material was placed on the pump chamber part to keep the part from being covered by PDMS solution because an actuating diaphragm should be fabricated on the pump chamber. Bubbles in PDMS were eliminated in vacuum, and PDMS was cured at 70°C for 2 hours. Then the cylinder was removed and refilled with PDMS solution in the mixing ratio of 20 part base to 1 part curing agent and a NdFeB magnet. Diameter of the magnet is 3.0 mm and thickness is 1.5 mm.
After curing for 2 hours at 70°C, whole PDMS was pulled apart from the wafer and holes for connecting tubes in inlet and outlet were punched. Polyethylene tubes (INTRAMEDIC®, inner diameter is 0.58mm) were connected to the reservoirs in each side of micropump. Finally, PDMS that had diffuser, nozzle and magnet-fixed chamber was attached to a slide glass. The fabricated micropump is shown in Figure 3.3.

With respect to the adhesion of cured PDMS to rigid smooth surface, it is possible to explain that the surface forces arise from (1) van der Waals forces, (2) elastomstatic forces and (3) hydrogen bonds. It is not clear which forces provide the major source of bonding, but most observations are in favor of the van der Waals forces. Lotters et al. [1997] reported that adhesive strengths of PDMS to polished tungsten were observed up to 180 kPa. If irreversible adhesion is necessary, tesla coil treatment is available.
Figure 3.1 Negative mask of the micropump. The middle circle is pump chamber (diameter 10mm) and two small circles in each side are reservoirs (diameter 3 mm) for connecting inlet- and outlet-tubes. The dimensions of diffuser and nozzle are described in Table 4.1 of chapter 4.
Figure 3.2 Fabrication of micropump using PDMS. The procedure after patterning on Si wafer is illustrated.
Figure 3.3  Real picture of micropump before connecting tubes.
3.2. Performance test of micropump

The experimental setup for flow measurement is illustrated schematically in Figure 3.5. This is similar to the method reported previously [Lee et al., 2004]. The micropump was placed on the magnetic stirrer as Figure 3.4 shows and two tubes from the inlet and the outlet were positioned horizontally on either side of the micropump. The end of inlet tube was immersed in the reservoir of working fluid. Before starting actuation, the inlet tube and the micropump were fully filled with the working fluid and the outlet tube was partly filled. The fluid head in the outlet was monitored during actuation. Back pressure of the micropump was controlled with changing the height of fluid surface on the reservoir.

The investigation for design of diffuser elements and characterization of the fabricated micropump were carried out. The specific parameters for the each experiment are described in chapter 4 with their results. If there is no special comment, the working fluid is methanol, and the back pressure is given as 78 Pa. And the frequency is 1 Hz.
Figure 3.4  Real picture of micropump actuation by a magnetic stirrer.

Figure 3.5  Schematic diagram of experimental setup for flow measurement
4. Results and discussion

4.1. Fabrication of diffuser/nozzle-based micropump

4.1.1. Design of diffuser elements

The pump chamber consisted of a circular cylindrical chamber (diameter: 10mm, thickness: 80μm). A diffuser and a nozzle were placed at each side of the chamber. The diffuser and the nozzle are key elements in the valveless pump and they have same shapes.

Diffuser efficiency is normally related to the Reynolds number and to three geometrical parameters, which are illustrated in Figure 2.6; neck width $W_1$, divergence angle $\alpha$, and slenderness $L/W_1$ (ratio of diffuser length to neck width). The micropumps with various diffuser length, neck width, and divergence angle were tested. The diffuser dimensions of the tested pumps are summarized in Table 4.1. The diffuser depth is 80μL. Dependence on each geometrical parameter was examined by flowing methanol and measuring pumping rate.

At first, three kinds of micropumps with different neck widths were tested and the pump performance is shown in Figure 4.1. Even though the neck width was increased to even and four times of the minimum one, the pump rate doesn’t change so much. The reason is easily understood with the equation (8). The effective parameter for changing diffuser-nozzle efficiency is the ratio of $A_1$ and $A_2$ rather than $A_1$ itself. Therefore, if $L/W_1$ and $\alpha$ remain constant, the ratio of $A_1$ and $A_2$ is constant and there is no significant change in the pump efficiency.

Meanwhile, the variation of length of diffuser can change the pump efficiency. Using the equation (4), (8), (14), and (15) the theoretical rectification efficiency is estimated in Figure 4.2. It increases proportionally at the low range of diffuser length. It is because too short
diffuser channel is not able to induce sufficient pressure drop for pumping. And rectification efficiency goes stable as the length of diffuser increases. Experimental data are plotted in Figure 4.3, which shows similar curve with estimated data at short length, while the pump rate decreases with the length increasing. It is due to error of theoretical assumption. When the ratio of length to neck width increases, $W_2$ becomes much larger than $W_1$ and lubrication approximation is no more available.

Concerning the divergence angle, relatively many researches have been reported. Depending on the fabrication constraints, the two operation ranges for a divergence angle are $5^\circ < \alpha < 10^\circ$ and $\alpha \approx 70^\circ$. At the range of $5^\circ < \alpha < 10^\circ$, theoretical rectification efficiency toward positive direction is maximized. On the other hand, the one toward negative direction is maximized at $\alpha \approx 70^\circ$ as Figure 4.4 shows [Nguyen and Wereley, 2002]. And Olsson et al. [1997] reported that the piezoelectrically actuated micropump, with a neck cross section of $80\times80$ $\mu$m, a diffuser angle $7^\circ$, and a length of $1093$ $\mu$m (ratio of the length to the neck width is 13.7), showed best pump performance. Here, the micropumps with $5^\circ$, $7^\circ$, and $9^\circ$ of divergence angle were examined and the results are shown in Figure 4.5. The maximum pump rate is obtained at $7^\circ$ of divergence angle.
Table 4.1  Pump dimensions and measured volumetric pump rate.

<table>
<thead>
<tr>
<th>Pump</th>
<th>$W_1$ (µm)</th>
<th>L (µm)</th>
<th>L/$W_1$</th>
<th>$\alpha$</th>
<th>Q (µL/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>100</td>
<td>1370</td>
<td>13.7</td>
<td>7°</td>
<td>12.80</td>
</tr>
<tr>
<td>2*</td>
<td>200</td>
<td>2740</td>
<td>13.7</td>
<td>7°</td>
<td>13.55</td>
</tr>
<tr>
<td>3</td>
<td>400</td>
<td>5480</td>
<td>13.7</td>
<td>7°</td>
<td>15.38</td>
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<tr>
<td>4</td>
<td>200</td>
<td>1000</td>
<td>5</td>
<td>7°</td>
<td>18.62</td>
</tr>
<tr>
<td>5</td>
<td>200</td>
<td>2000</td>
<td>10</td>
<td>7°</td>
<td>18.91</td>
</tr>
<tr>
<td>6</td>
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<td>3600</td>
<td>18</td>
<td>7°</td>
<td>10.20</td>
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<td>13.7</td>
<td>5°</td>
<td>7.24</td>
</tr>
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<td>200</td>
<td>2740</td>
<td>13.7</td>
<td>9°</td>
<td>12.84</td>
</tr>
</tbody>
</table>

* Dimensions of this micropump were utilized as fundamental values in the other experiments.
Figure 4.1  Pump rate depending on the neck width of diffuser element. The data of pump 1, 2, and 3 on Table 4.1 are plotted (L/W1=13.7 and α=7°).
Figure 4.2  Estimated rectification efficiency depending on the ratio of length to neck width when $\alpha=7^\circ$. 
**Figure 4.3** Pump rate depending on the length of diffuser element when $W_1=200 \ \mu m$ and $\alpha=7^\circ$. The data of pump 2, 4, 5, and 6 on Table 4.1 are plotted. The point at $L=0$ is obtained theoretically.
Figure 4.4 Theoretical function of rectification efficiency and experimental data. Here $\theta$ denotes the divergence angle, $\alpha$ [Nguyen and Wereley, 2002].
Figure 4.5  Pump rate depending on the divergence angle of diffuser element. The data of pump 2, 7, and 8 on Table 4.1 are plotted ($W_1=200 \, \mu m$ and $L/W_1=13.7$).
4.1.2. Diaphragm of micropump

Dow Corning silicone elastomer products are supplied in two parts as lot-matched base and curing agent that are mixed in a ratio of 10 parts base to one part curing agent by weight. Since curing agent contains cross-linker, the mixing ratio of two parts is important factor for determining cross-linking density of cured PDMS.

In this experiment, the frame of micropump and the diaphragm were made in different mixing ratios of the base to the curing agent. The mixing ratio for the frame was 10 to 1 as the product manual standardizes. And the diaphragm contains less curing agent than the frame has, in a ratio of 20 parts base to 1 parts curing agent.

The mechanical properties of cured PDMS in different ratios were measured using INSTRON universal testing machine (static 5583). The samples for testing were made by cutting the cured PDMS plate. The width, thickness, and spec gauge length of samples are 6.25mm, 1.5 mm, and 25mm, respectively. The mixing ratios of samples were 10 to 1 and 20 to 1. Four samples per each composition were fabricated and the results were averaged. As results are shown in Table 4.2, the diaphragm of mixing ratio of 20:1 has more elastomeric property than the one of mixing ratio of 10:1. Two micropumps with the diaphragms which have different PDMS-mixing ratio were fabricated and their pumping performance was tested. And it was founded that the pump rate was enhanced as mixing ratio was increased because oscillation of the more elastic diaphragm was carried out more easily.
Figure 4.6  Tensile properties of PDMS samples which were cured in two mixing ratio.

Table 4.2  Mechanical properties of PDMS in different ratios of base to curing agent.

<table>
<thead>
<tr>
<th>Mixing ratio (Base : Curing agent)</th>
<th>Displacement at break (mm)</th>
<th>% Strain at break (%)</th>
<th>Load at break (kN)</th>
<th>Stress at break (MPa)</th>
<th>Young’s modulus (MPa)</th>
<th>Pump rate (μL/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10:1</td>
<td>67.83</td>
<td>271.3</td>
<td>0.0398</td>
<td>4.486</td>
<td>1.662</td>
<td>7.25</td>
</tr>
<tr>
<td>20:1</td>
<td>70.11</td>
<td>280.5</td>
<td>0.0091</td>
<td>1.207</td>
<td>0.428</td>
<td>13.55</td>
</tr>
</tbody>
</table>
4.2. Performance test of micropump

4.2.1. Characterization of reciprocating fluidic motion

The flow induced by the diaphragm pump shows the oscillation as explained previously. Since the micropump in this experiment was actuated in relatively low frequency range, the oscillating flow of working fluid was able to be observed. The movement of flow head in outlet tube was monitored using a video camera and illustrated in Figure 4.7. The positive sloped parts represent pump modes and the negative sloped parts do supply modes. The maximum and minimum points indicate the transitions between the supply and the pump modes.

The micropump was also examined for frequency dependence. Methanol was used as working fluid. In order to neglect capillary effect, the back pressure of 100 Pa was given to the device. With respect to the relation between the stroke volume and frequency, it is obvious that the deflection of diaphragm decreases with frequency increasing. At low frequency, the diaphragm has enough time to bend and the large stroke volume can be obtained. The pumped volume per one complete cycle is also relatively large. However, as the frequency increases, the movement of diaphragm becomes much faster and the stroke volume decreases along with the pumped volume for net flow. The stroke volume and compression ratio $\varepsilon$ are depicted in Figure 4.8. According to the experimental data, the stroke volume and the compression ratio at 0.91 Hz are 728 nL and 11.6 %, respectively. Meanwhile, those values at 7.5 Hz were measured as 13.5 nL and 1.81 %, respectively. Frequency can be thought to affect a lot for reciprocating motion of fluid.

However, we should notice that the percentage of net pumped volume out of the stroke volume does not decrease as much as the stroke volume does when the frequency increases. In other words, when frequency increases, the stroke volume and the net pumped volume per
one stroke (or also called 'cycle') decrease simultaneously. Thus, the rectification efficiency decreased slightly with respect to change of frequency as shown in Figure 4.9. The ratio of pressure-loss coefficients of nozzle and diffuser was also calculated with the equation (4). Here it is confirmed that the $n_{nd}$ is the parameter depending on the geometry of micropump rather than stroke variation of pump. $n_{nd}$ of this micropump is about 1.37 as shown in Figure 4.10.
Figure 4.7  Reciprocating movement of fluid in outlet tube. $\Delta L$ is defined as the distance from the outlet hole of micropump to the fluid head in the tube as shown in Figure 3.5. $t$ is actuation time. Reciprocating flow was monitored at the frequency of (a) 0.909 Hz, (b) 1.90 Hz, and (c) 4.11 Hz.
Figure 4.8  Stroke volume (left axis) and compression ratio (right axis) depending on frequency.
Figure 4.9  Rectification efficiency $\mathcal{X}$ depending on frequency.
Figure 4.10  Ratio of pressure-loss coefficients of nozzle and diffuser.
4.2.2. Characterization of the net flow

In Figure 4.7, we can see the maximum (or minimum) value of $\Delta L$ increase gradually and it indicates that the net flow is induced toward positive direction. The net pump rate was obtained by plotting the maximum (or minimum) values. According to the literature, various micropumps have shown a tendency of having maximum pump rate in specific frequency and decreasing pump rate as frequency increases [Koch et al., 2000; Woias, 2004]. In Figure 4.11 the maximum pump rate is observed around 3 Hz. However it seems to be more natural to analyze that the pump rate is stable over the tested frequency range except the last one. The reason of stable tendency can be explained as follows. As discussed in the previous section, with higher frequency the smaller volume was pumped for one cycle. Simply speaking, relatively large volume of fluid is transported a few times at low frequency while small volume of fluid is transported more times at higher frequency. The frequency times the corresponding pumped volume per one cycle makes similar results along the variation of frequency. Thus the net pump rates become similar. Additionally, the tested range of frequency here may be not large enough to search the tendency of pump rate depending on frequency. The micropump is expected to have lower rectification efficiency at higher frequency and the pump rate would be decreased because the rectification efficiency is decreasing slowly with frequency increasing as shown previously in Fig 4.10.

The behavior of micropump with respect to back pressure was also investigated. The working fluid was methanol and the frequency was 1 Hz. An almost linear drop of the pump rate was observed when increasing the back pressure as shown in Figure 4.12. The maximum pump rate was 18.2 $\mu$L/min with no back pressure. When the back pressure is 156 Pa, net flow would not be observed because the positive and negative flows in the pump chamber are same.
Figure 4.11  Net pump rate depending on frequency.
Figure 4.12  Net pump rate depending on back pressure.
4.3. Pumping properties for flowing various fluid

It is known to be one of the most dominant advantages of displacement micropumps that they are insensitive to fluids unlike dynamic micropumps. The displacement micropump is not sensitive to the ionic strength, or pH of the pumped media. However pump performance can be changed in viscosity variation due to adhesion forces [Andersson et al, 2001]. To examine that, methanol, ethanol, and isopropyl alcohol were utilized. These three alcohols have very similar properties such as density and surface tension, but have different viscosities, which are 0.5555 cP for methanol, 1.1474 cP for ethanol, and 2.32 cP for isopropyl alcohol. Figure 4.13 shows the back pressure dependency of each fluid at the frequency of 1 Hz. All fluids shows linear drop with increasing the back pressure, but the pump rates are quite different for each fluid.

The maximum pump rate vs. viscosity is plotted in Figure 4.14. Methanol which has smallest viscosity among the three working fluids showed in the highest pump rate. The decreasing tendency of pumping effect as increasing viscosity in this experiment is agreed with the estimated results. If the working fluid is assumed to be incompressible at all and the velocity of fluidic head in outlet tube can be converted to the flow rate in diffuser element directly, Reynolds numbers in this experiment are obtained as 34.58 for methanol, 8.35 for ethanol, and 2.74 for isopropyl alcohol.
Figure 4.13  Pumping performance using various working fluids.
Figure 4.14 Pump performance depending on viscosity at zero back pressure. The dots indicate measured pump rate (left axis) and the line indicates estimated rectification efficiency (right axis) which is calculated in chapter 2.2.1.
4.4. Effect of the external magnet rotation

Utilization of an external magnet for actuation of microdevices eliminates complex wiring between the microdevice and outer facilities and makes the microdevice simple. However it should be noticed that the movement of diaphragm by the permanent magnet is particular a little. The small magnet-embedded diaphragm shows not only vertical displacement but also distortion due to lateral rotation of the magnet. As Figure 4.15 (a) shows, the diaphragm happens to be distorted in the junction of external magnetic field. The schematic diagrams of the micropump and the external magnet of the stirrer are drawn sequentially from (b) to (e) in Figure 4.15. Two different magnetic fields are indicated by the dotted box and the filled box. The chamber cavity is grey-colored. Since the scale of micropump is so small, such a movement may affect fluidic flow in the pump chamber. So the pump rate was measured with changing the rotating direction of external magnet. Methanol was used for working fluid. The rotating directions are illustrated in Figure 4.16. As a result, the pump rate is the highest when the external magnet rotates in counter-parallel direction to the fluidic transport. And co-parallel directed rotation is revealed to produce the lowest pump performance (Figure 4.17). If the external magnet rotates in counter-parallel direction, it is expected that the fluid near the outlet of pump chamber is squeezed prior to the fluid near the inlet in pump mode. Such a partially different fluidic behavior in the pump chamber results in enhancement of fluidic transport toward the outlet. On the other hand, the fluid near the inlet is squeezed prior to the fluid near the inlet in co-parallel directed rotation. So, the backward flow to the inlet increases in pump mode. Whether the diaphragm is distorted or not, the major movement of the diaphragm is made vertically like other actuating methods. Thus, fluid transport was also observed with rotating the magnet in perpendicular direction.
Figure 4.15  Schematic diagrams for the movement of diaphragm by the external magnet which rotates clockwise from top view.
Figure 4.16 Relative rotating directions of the external magnet to the micropump
Figure 4.17  Pump rate depending on rotating direction of the external magnet. (a), (b), (c) and (d) are defined in Figure 4.16.
4.5 Durability of PDMS diaphragm

To verify compatibility of PDMS as a diaphragm material, endurable pump performance is important. So the pump rate for methanol was measured at 1Hz over one hour continuously. As Figure 4.18 shows, the initial pump property was lasted over 20 minutes although the pump rate slightly decreases later due to PDMS swelling by methanol. If the micropump would be utilized for micro total analysis systems, working period should not exceed over several minutes. So that PDMS can have acceptable durability.

![Graph showing pump rate vs duration time](Image)

**Figure 4.18** Change of pump performance during experiment.
5. Conclusions

Diffuser/nozzle-based micropump was fabricated by simple method using polydimethylsiloxane elastomer. A NdFeB magnet on the diaphragm of micropump is vibrated by periodical change of magnetic force. The magnetically actuated micropump showed specific dependence with regard to the diffuser dimensions such as neck width, length of diffuser, and divergence angle.

Two kinds of diaphragms were made in variation of PDMS mixing ratio. The diaphragms showed different mechanical strengths affecting pumping performance. Thus, differing PDMS mixing ratio of diaphragm makes it possible to fabricate various micropumps, which have same dimensions but show different pumping performances. In other words, only one template of micropump is necessary and photolithographic steps after first fabrication can be eliminated.

The micropump induced reciprocating fluidic motion. As frequency increases, the stroke volume and pumped volume decrease. The rectification efficiency was around 8% at frequency of 1Hz and the pump efficiency was almost constant over the tested frequency range. The frequency and back pressure dependencies of the micropump were investigated. The maximum pump rate at 1 Hz and zero back pressure was achieved as 18.2 µL/min. And methanol, ethanol, isopropyl alcohol were used as working fluids. The higher viscosity the fluid has, the lower pump rate was observed.

Unlike the other actuating methods, magnetic actuation with rotating permanent magnet induced the distortion of the diaphragm. Therefore the pump rates were different depending on the rotating direction of the external magnet. When the flow direction and the magnet rotating direction are counter-parallel, best pump performance was observed.
Finally, durability of PDMS diaphragm was tested. The initial pump performance lasted over 20 minutes although swelling of PDMS occurred slowly.

Easiness of fabrication and operation is obviously the big advantage of this pump. There are no valves, no electrodes and no physical connection between outer actuator and micropump. The structure is extremely simplified, although the micropump shows effective pumping performance relatively. However, reciprocating micropumps are susceptible to problems such as self-priming and bubble tolerance. Since the scale is very small, tiny particle or small gas pockets trapped in the chamber can affect the results severely. To overcome such problems, investigation about robust design is necessary.
국문 요약

본 연구에서는 자기력에 의해 구동되는 미세유체펌프의 제작 방법을 모색하고 그 유체 전달 특성을 분석하였다. 이 펌프는 브러스 없는 디퓨저 노즐 타입으로서 소프트 리소그래피를 이용하여 poly dimethylsiloxane(PDMS)으로 제작하였다. 펌프의 다이아프램에 NdFeB 자석을 고정하여서 외부 자기력에 의해 다이아프램이 수직운동을 하도록 설계하였으며, 외부 자력원으로는 교반기가 이용되었다.

먼저 다양한 디자인(폭 너비, 길이, 발산각수 변화시킴)의 미세펌프를 제작한 후 각각의 경우 유속을 측정함으로써, 디퓨저의 구조적인 특성이 펌프 특성에 미치는 영향을 실험적으로 분석하였다. 그리고 PDMS의 배합비율을 다르게 해서 다이아프램의 기계적 탐성력에 따른 펌프 특성의 변화를 관찰하였다.

주파수와 후방압력에 따른 펌프 효과를 살펴보면, 주파수가 증가할수록 스트로크 부피와 한 스트로크 당 전달된 유체의 양은 감소하나, 펌프 효율은 일정함을 보였다. 그리고 후방압력의 증가에 따라 유속은 선형적으로 감소했다. 펌프 유체로는 메탄올과 에탄올, 이소프로필알콜을 사용하여 유속을 측정하였는데, 점도가 증가함에 따라 펌프 유속이 감소함을 확인할 수 있었다.

그리고 교반기의 자석 회전과 미세유체펌프 내의 유체 순환 간의 상대적인 방향성에 의한 펌프 특성에 관해 고찰하였다.

마지막으로 메탄올에 의한 PDMS의 방장으로 다이아프램의 기계적 특성 변화가 생기는지를 살펴보기 위해 장시간 연속적으로 구동시킨 결과, 초기 유속이 20분 이상 지속됨이 확인되었다.
References


감사의 글

2년이라는 기간 동안 배우고 얻었던 저의 미력한 성과가 이렇게 논문으로 정리되기까지 많은 분들의 도움 없이는 불가능 했을 것입니다. 이에 이 편을 빌려 무한한 감사의 말씀을 전합니다.

항상 학자로써의 모범을 보여주시고 저의 연구 방향을 인지하게 이끌어주신 김도현 교수님께 감사 드립니다. 그리고 바쁘신 와중에 저의 논문 연구와 심사에 큰 도움을 주신 양승만 교수님과 박오옥 교수님께 감사 드립니다.

학문 뿐만 아니라 인생의 조언자로 도움을 주신 선혜 오빠, 저의 실험 분석에 애정 어린 조언을 아낌없이 주셨던 승엽 오빠, 이론적인 해석에 많은 가르침을 주신 영수 오빠, 언제나 따뜻한 관심으로 다독여주신 재신 언니, 많은 아이디어를 알려주신 실험적 기술을 가르쳐 주신 광석 오빠, 연구실 생활이 늘 원만하도록 배려해주신 고도와주셨던 범승 오빠와 영훈 오빠, 그리고 어린 선배 밑에서 실험실의 많은 일들을 하느라 수고하시고 협이 되어주시는 석환 오빠, 유순 오빠, 일관에게 감사 드리며 건강과 행운을 기원합니다.

그리고 항상 저의 고민 상담을 들어주시고 격려해주시는 최고의 콜메이트 지영 언니와, 오랜 사귀므로 이제는 형제 차례와 같이 의지가 되는 고등학교 동기 여학생들, 풍물根底 소리모음 사람들이라는 저의 카이스트 생활이 즐거울 수 있었습니다.

마지막으로, 힘든 기간 동안 제가 나아갈 수 있도록 든든한 비밀목이 되어 주신 저의 가족과 정무 오빠에게 온 마음을 다해 사랑과 감사의 말씀을 드립니다.
이력서

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