

A Novel Micromachined Magnetic Membrane Microfluid Pump

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Abstract – We present results on the design, fabrication, and testing of a new, micromachined magnetic pump for integrated microfluidic systems. Structurally, the pump consists of a magnetic microactuator and two polymer-based one-way diffuser valves. The microactuator is based on a thin membrane made of polydimethyl siloxane (PDMS), a soft silicone elastomer. Membrane displacement is caused by the interaction between ferromagnetic pieces (embedded within the thickness of the membrane) and an external magnet. This novel mechanism reduces fabrication and packaging complexity, and allows for remote operation of the micropump without any tether wires for power input. The operation is simple as no precise alignment is required between the external magnet and the pump. PDMS is bio-compatible compared to silicon. One future application of this tetherless micropump is implanted biomedical microfluidic systems.

We developed a novel micromachining process for embedding ferromagnetic materials (Permalloy, $\text{Ni}_{80}\text{Fe}_{20}$) within a thin, spin-casted PDMS membrane. Unique pump and diffuser mechanisms that allow for continuous pumping are also developed. Diffuser elements containing no moving parts are fabricated using polymer micromachining techniques. Micro Permalloy pieces are strategically positioned within a $2 \times 2 \text{ mm}^2$, $40\text{-}\mu\text{m}$ -thick PDMS membrane. Dimensions and locations of the membrane and the Permalloy pieces are optimized using computer simulations for maximum membrane vertical displacement under a given magnetic field. Experimentally, we have demonstrated successful on-chip fluid pumping. In the presence of an oscillating 2.85×10^5 -A/m external magnetic field, a $1.2\text{-}\mu\text{l}/\text{min}$ flow rate was measured for an actuation frequency of 2.9-Hz. The flow rate can be easily varied by the frequency.

Key words – micropump, microfluidic system, membrane, magnetic actuation, silicone elastomer.

I. INTRODUCTION

Micro liquid handling systems [1] has been an area of increasing interest in the field of MEMS during recent years. Mechanical microfluidic handling systems consist of micropumps and microvalves, which employ various actuation mechanisms [2–6]. The actuation principles that have been applied to membrane micropumps include piezoelectric [2], electrostatic [3], thermopneumatic [4], bimetallic [5], and electromagnetic [6]. The membrane material chosen for these devices is silicon [2–6]. The types of microvalves and flow controllers used include passive check valves [3, 5], active diaphragm valves [6], and nozzle-diffuser pairs [2, 4]. For many electrostatic and piezoelectric

membrane actuators, large actuation voltages is needed. For example, the PZT actuator for a micropump discussed by Koch *et al.* [2] needed a 600 V_{pp} driving voltage.

Since the achievable displacement for flat-membrane actuators is generally limited (a few microns to $10\text{--}20\mu\text{m}$), the overall volume flow rate of micro membrane pumps is thereby limited as well. To achieve larger deflections, novel structured membranes such as corrugated membranes [4] have been fabricated, although they require a more involved fabrication process. Jeong *et al.* [4] compared deflections for flat and corrugated silicon diaphragms of the same dimensions. A $4 \times 4 \text{ mm}^2$ membrane with 7 corrugation rings deflected by $37.5\text{-}\mu\text{m}$ while a flat membrane only yielded a $11.7\text{-}\mu\text{m}$ deflection under a 6V applied voltage. Alternately, elastic materials such as silicone elastomers [11] are used for their low Young's Modulus. This means that larger membrane displacements is achievable with similar power inputs. Larger displacements in the actuator translates into larger stroke volumes and higher flow rates in the micropump. Besides their favorable mechanical properties, silicone elastomers are physically and chemically stable and inert, thus making them bio-compatible.

Magnetic actuation [6–8] has been explored for its favorable characteristics. It is shown to produce large forces (few hundred μN), which affect large displacements. Most of these micropumps and microvalves are driven by integrated magnetization sources, which require wire feed. A group that explored magnetic effects as an actuation principle, Zhang *et al.* [6], fabricated a magnetic membrane micropump with a $7\text{-}\mu\text{m}$ thick Permalloy film on a $17\text{-}\mu\text{m}$ -thick, $8 \times 8 \text{ mm}^2$ silicon membrane. This membrane achieved $23\text{-}\mu\text{m}$ deflections when driven by integrated inductors operating at 300 mA DC and 3 V .

The novel aspect of this work is the fabrication of the first tetherless micromachined membrane pump, based on polymer material and external magnetic actuation. In comparison, our membrane is four times smaller than the one reported in [6] and achieves four times greater displacements. Since our actuation force is provided by an external magnetic field (for which great flexibility is given to its positioning), it can be remotely operated without needing any wires for power input to the device. It is known that material damage is not a concern under high magnetic fields up to 1 Tesla [7]. Testing of our device was done at low fields of 0.11 to 0.23 Tesla, which is sufficient in producing the large displacements and thus the flow rates measured.

II. PRINCIPLE AND DESIGN OF THE PUMP

The pump consists of a magnetic membrane actuator (see II in Fig. 1). When actuated by a magnetic field, its membrane deflects and pushes fluid out of the pump chamber. The

direction of fluid flow is then controlled by one-way diffusers placed at the inlet and outlet of the pump chamber.

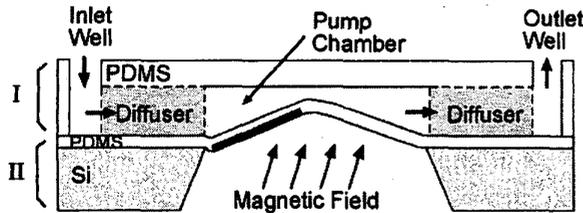


FIGURE 1. Cross section of assembled micropump.

(A) MEMBRANE ACTUATOR

The structure of the magnetic actuator is shown in Figure 2. A thin layer of PDMS rests on the front surface of a silicon wafer, through which a square through-hole has been etched. Rectangular pieces of Permalloy are embedded within the PDMS membrane. This array of flaps is arranged parallel to each other only along one side of the membrane.

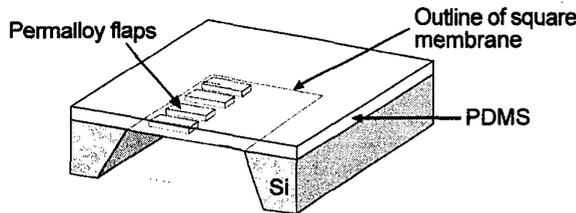


FIGURE 2. Schematic cut-out illustration of the membrane actuator (cut across its symmetry plane to illustrate the cross-section).

Figure 3 illustrates the actuation principle of the magnetic membrane actuator. The default mode for the membrane, the Rest Mode, occurs under zero magnetic field. In the presence of an external magnetic field (provided by a permanent magnet or an electromagnet), a torque is generated that causes the Permalloy flaps to deflect. As the flaps are deflected, they displace the membrane, thereby causing the movement shown in the Actuation Mode of Figure 3. With this, a net volume displacement is produced.

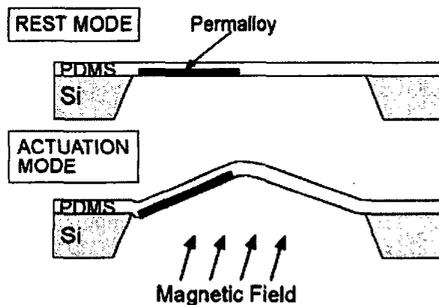


FIGURE 3. Actuation principle of the magnetic membrane actuator.

To apply the actuator to a pump, it is highly desired that the overall volume displacement be as large as possible under a given magnetic field and membrane dimensions. Thus, several design issues need to be addressed. Key design

parameters include: (1) the length, width, and height of the Permalloy pieces, (2) the number of flaps, (3) the size and thickness of the membrane, (4) the spacing between Permalloy pieces, and (5) the spacing between the Permalloy pieces to the edge of the membrane (parameter A in Fig. 4).

It is known that the magnitude of the magnetic torque is generally proportional to the volume of the Permalloy piece [7]. Thus, larger torques could be achieved with longer Permalloy pieces. However, the membrane becomes stiffer, thus limiting its flexibility to stretch and deflect. On the contrary, if the flaps are short, the membrane will be more flexible but the actuation torque will be smaller. A similar consideration must be assigned in determining flap width and thickness, as well as flap spacing and placement. To satisfy the actuator's design requirements, computer simulation is used to optimize membrane displacement by varying the key design parameters. Finite element analysis (ANSYS) results yielded the design layout shown in Fig. 4.

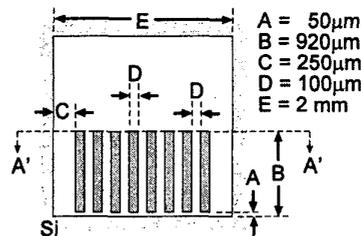


FIGURE 4. Layout (top view) of Permalloy flaps in a 2x2 mm² membrane. The A'-A' line defines the cross-section at which membrane displacements will be measured and plotted.

(B) DIFFUSER ELEMENT

Design of our diffuser elements follow the design, simulation, and measurement studies by Olsson *et al.* [9-10]. These no-moving-parts diffuser elements have diverging walls in the positive flow direction (see Fig. 5), which promotes one-way fluid flow in the positive direction. For best pump efficiency, diffuser pressure losses in the negative direction need to be maximized while minimizing losses in the positive direction [9]. Diffusers used in this pump have a 10° opening angle, neck-widths of 100-μm and 500-μm, and is 2.3-mm long, as shown in Figure 5.

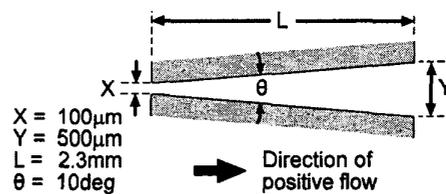


FIGURE 5. Layout of diffuser element with pertinent parameters.

III. FABRICATION

(A) MEMBRANE ACTUATOR

Major steps in our fabrication process are illustrated in Figure 6. A 4000-Å thick film of silicon dioxide is thermally grown on a <100> oriented silicon wafer. Identical alignment

marks are then patterned on the oxide on both the front and back surfaces. Windows are etched through the oxide on the back surface (Fig. 6a) for a targeted $2 \times 2 \text{ mm}^2$ window on the front surface after the wafer is etched through. Etching is performed in EDP silicon etchant (Transene PSE-300) at 100°C until a thin layer ($50\text{-}\mu\text{m}$) of silicon is left (Fig. 6b), which will provide structural support in further processing. Next, a $100\text{-}\text{\AA}$ Cr adhesion layer and $2000\text{-}\text{\AA}$ Cu seed layer (for electroplating) is evaporated onto the front surface (Fig. 6c). A $10\text{-}\mu\text{m}$ thick layer of photoresist AZ4620 is then spin-coated on the metals and lithographically patterned.

Permalloy material ($\text{Ni}_{80}\text{Fe}_{20}$) is electroplated to a total height of $22\text{-}\mu\text{m}$, over the photoresist mold height (Fig. 6c). This over-electroplating is intentional as it causes lateral Permalloy electroplating over the PR mold edges, in addition to the typical vertical electroplating. The shape of this over-electroplating is confirmed in Figure 7. The lateral extension provides crucial physical restraint to secure the Permalloy flaps within the membrane when it is actuated. The Cu and Cr are then etched. PDMS (DuPont Sylgard 184) mixed at a 1:10 ratio of curing agent to elastomer base, is poured onto the front surface of the wafer and allowed to flow over and around the Permalloy structures. The wafer is then put into a vacuum chamber to remove any air trapped underneath the lateral extension. It is then spun to achieve a $40\text{-}\mu\text{m}$ PDMS layer (Fig. 6d). This layer is allowed to reflow under room temperature to produce a planar top surface and then cured in a 100°C convection oven for 15 minutes.

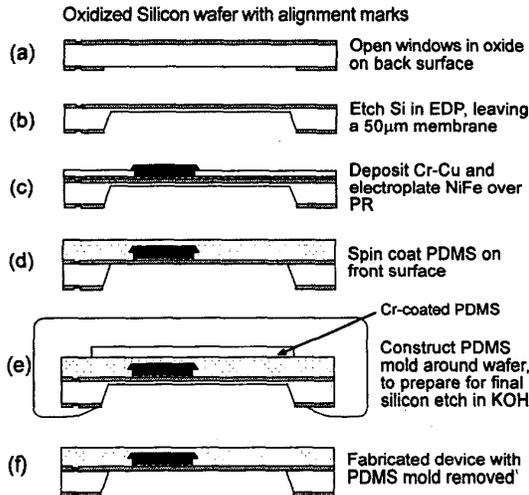


FIGURE 6. Fabrication process for magnetic membrane actuator.

As a final step, removal of the remaining $50\text{-}\mu\text{m}$ silicon layer is needed to release the PDMS membrane. As the etch solution could diffuse through the PDMS membrane, directly immersing the wafer into an anisotropic wet etchant is undesirable. To prevent this, we construct a PDMS mold around the wafer, leaving only the back window open (Fig. 6e). Before this, a 1-mm thick PDMS piece, large enough to cover the membrane and double-side coated with $540\text{-}\text{\AA}$ -thick sputtered Cr, is placed on top of the membrane. This Cr-

coated piece eliminates the potential contact between the PDMS mold and the PDMS membrane. Without this piece, strong PDMS-PDMS adhesion will cause sticking and result in membrane peel-off during mold removal. The protective mold is then constructed and cured in a 100°C oven for 30 minutes. The final silicon etch is completed in KOH (33% by weight in DI water). KOH is chosen as the final etchant as the mold remains transparent, which eases its removal. After the window is etched through, the protective mold is carefully removed to release the device (Fig. 6f).

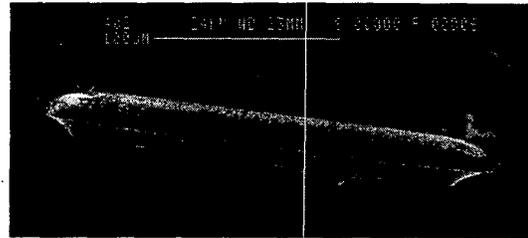


FIGURE 7. SEM micrograph showing shape at sides of over-electroplated Permalloy flap.

(B) Diffuser Element, Channels, and Pump Chamber

The inverse pattern of the diffuser elements, pump chamber, inlet and outlet wells, and microchannels ($300\text{-}\mu\text{m}$ wide), are designed on the same mask. This mask is then used to photolithographically pattern a $260\text{-}\mu\text{m}$ thick layer of SU-8 epoxy-based negative photoresist. PDMS is then poured over the SU-8 structures and cured in a 100°C oven. The PDMS layer is then peeled off the SU-8 structures and the desired components are thus molded into the PDMS layer.

(C) Micropump Assembly

The fully assembled micropump is shown in Figure 1. With pump chamber to actuator window alignment, the two layers are bonded by lightly pressing the PDMS diffuser layer (I) onto the PDMS membrane actuator (II). Strong PDMS-PDMS adhesion thus provides sufficient bonding.

IV. MEASUREMENTS AND RESULTS

(A) MEMBRANE ACTUATOR

The experimental setup used for membrane actuator testing is shown in Figure 8. Our goal is to determine a relationship between membrane displacement and applied magnetic field. The actuation source is a Neodymium-Iron-Boron (Edmund Scientific NdFeB 27/30) magnet with a surface magnetic field strength of $2.14 \times 10^5 \text{ A/m}$. Deflections are measured using an optical microscope. An objective lens is used to focus onto the same corner, in the presence and absence of the magnet, and the needed change in lens re-focusing is converted to a finite height difference. The applied magnetic field is determined by measuring the distance D in each case, and can be easily varied by D .

The minimum magnetic field strength needed to initiate observable membrane displacement is $3.18 \times 10^4 \text{ A/m}$. Figure 9 plots average displacements attained from successive measurements performed on the actuator at magnetic field

strengths of 1.85×10^5 -A/m, 1.38×10^5 -A/m, and 8.91×10^4 -A/m, resulting in average maximum deflections of 83.9- μ m, 67.9- μ m, and 39.6- μ m, respectively.

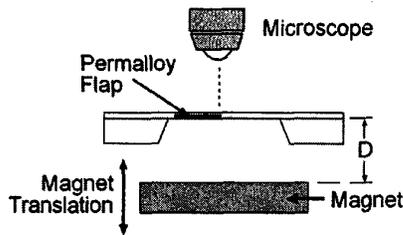


FIGURE 8. Schematic of the testing setup, measured by focusing on a corner of each Permalloy flap for each given magnetic field.

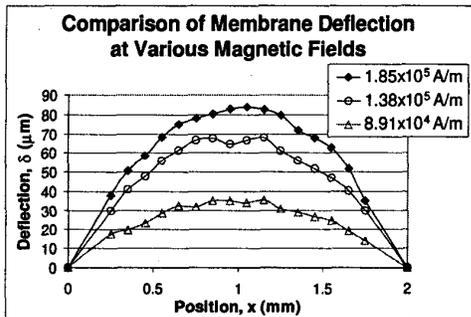


FIGURE 9. Plot shows average membrane displacements along the A'-A' cross-section in Figure 4, tested at applied magnetic fields of 1.85×10^5 , 1.38×10^5 , and 8.91×10^4 -A/m. The points represent the displacements measured at each corner of each Permalloy flap.

(B) MICROPUMP

Flow rate measurements were recorded using the test setup shown in Figure 10. Undiluted food dye is placed over the inlet well and driven by an oscillating 2.85×10^5 -A/m external magnetic field. Flow measurements were recorded by measuring the time the dye takes to travel a particular distance through the microchannels. A flow rate of 1.2- μ l/min was measured for a 2.9-Hz pumping frequency.



FIGURE 10. Top view of testing setup for flow rate measurement.

V. CONCLUSIONS

This paper reports the development of a unique, micro magnetic membrane pump, which has been successfully fabricated and tested. A novel fabrication process for embedding ferromagnetic pieces in PDMS is developed for the first time. The use of over-electroplated Permalloy flaps

embedded within the membrane, permits the use of an externally applied magnetic field for actuation. The membrane is made of a flexible silicone elastomer (PDMS), thus allowing for large vertical membrane displacements, which is desirable for micropumps. Greater stroke volume is possible with larger membrane displacements, which can be easily controlled by increasing the magnetic field strength. Higher flow rates can also be realized by increasing the actuation frequency. This micropump is characterized by numerous favorable attributes, including remote operation, simplicity of fabrication and operation, large achievable stroke volume, good fluid flow rates, safe and bio-compatible (PDMS is both physically and chemically stable), and long-term reliability. This type of micropump can be applied to implanted biomedical microfluidic systems.

VI. ACKNOWLEDGEMENT

This work is supported by DARPA Composite CAD program under contract number F 30602-98-2-0178.

VII. REFERENCES

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