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Design and Development of a PDMS Membrane based SU-8 Micropump for drug Delivery System

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Drug Delivery System, PDMS, SU-8, microfabrication, micropump, Piezoelectric

Abstract

A micropump is one of the key components of a Drug Delivery System (DDS) and it can be integrated with microneedle for localized drug delivery in even one single cell. The conventional drug delivery system is replaced by MEMS (Micro-electro-mechanical system) DDS due to its ability to deliver very small amounts of the drug in the dermis region of the patient's body without any pain. This paper presents the design and fabrication of a PDMS (Polydimethylsiloxane) membrane based SU-8 micropump. The pump chamber is made of SU-8 on a glass substrate. A separately processed PDMS layer of 150 μm is treated in oxygen plasma to make a strong bond with SU-8. The detailed fabrication procedure is described here and the advantage of using a PDMS membrane as well as SU-8 material is also discussed. Piezoelectric discs are used for the actuation of the micropump membrane. The deflections of the membrane at different frequencies are measured using MSA 400 (Micro System Analyzer) of Polytec and they are shown to be maximum at 5009 Hz frequency for 30 V. The flowrate measurement using DI water is also discussed here.

1. Introduction

Micro-electro-mechanical system (MEMS) based microfluidic devices have received attention in the biomedical field over the last decade. Several active research groups are presently involved in developing micropumps and microneedles separately, and with their integrated version for the precise control of drug delivery in a localised area or even in a single cell. The concept of an integrated drug delivery system is illustrated in figure1.

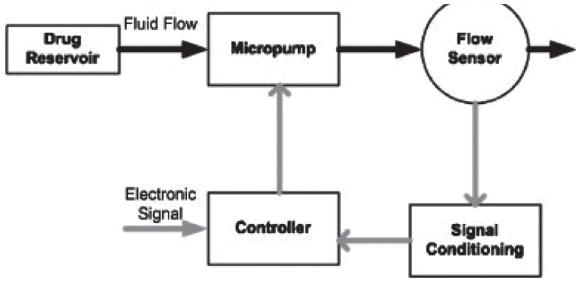


Fig. 1 A block diagram of an integrated microfluidic device for drug delivery

The development of the micropump based on microfabrication technology was started in the 1980s. The MEMS based micropump was first developed in the 1990s [Ashraf et al., 2011]. The micropump is the main component of a drug delivery system that provides the actuation mechanism to deliver specific volumes of therapeutic agents or drugs from a reservoir. The requirements for a drug delivery system include a minimum flow rate in the order of 10 μL per minute or more, small size and high reliability [Nisar et al., 2008]. Typically, a micropump consists of a diaphragm membrane, chamber, actuator, microchannels, microvalves, inlet, outlet, etc. Passive valves are more complicated in integrated microfluidic systems and they also suffer due to fatigue and clogging. Hence, the valveless diffuser micropump is more suitable for drug delivery.

An actuator is an essential and driving part of a micropump that converts energy into motion. It is used to provide force for fluid flow in micropumps. The actuator takes energy from electricity, heat, liquid pressure, and air pressure and converts it into motion. In most micropumps reported in the literature, the piezoelectric actuator disk is mounted with a membrane which pushes the fluid.

Chamber design is a very important and critical part in microfluidic systems and it can significantly influence the volume stroke, pressure characteristics and nozzle-diffuser loss coefficients. Most of the micropumps reported in the literature have a single chamber configuration. However, in order to improve the performance, two or three chamber micropumps have also been reported. Micropumps, in which pumping chambers are arranged sequentially or fabricated in such a way that the multiple chambers are in series or in parallel arrangements, are known as peristaltic micropumps.

2. Operating Principle

A schematic of the nozzle/diffuser action in micropumps is shown in Figure 2. The nozzle/diffuser element works in such a way that during the supply mode more fluid enters the chamber through an inlet than fluid that exits the outlet. In the pump mode, the reverse action occurs. Stemme and Stemme [1993] were the first to report valveless miniature micropumps in which they reported using a nozzle/diffuser element as a flow-rectifying element.

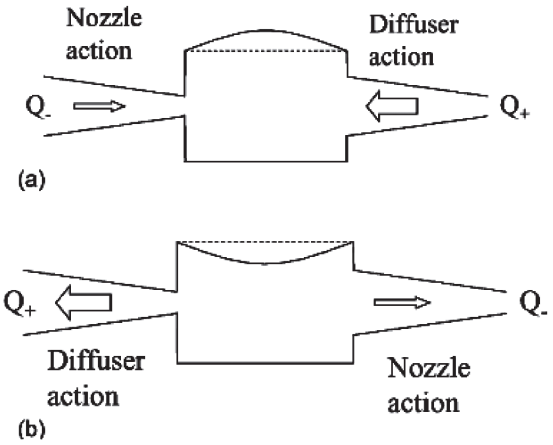


Fig. 2 Schematic of a valveless diffuser micropump (a) supply mode, (b) pump mode

The first piezoelectric micropump was fabricated using micromachining technology by Van Lintel et al. [1988]. The micropump consisted of a pumping chamber, a passive silicon (Si) check valve, and a thin glass membrane actuated by a piezo disk. A maximum flow rate of 8 μL/min and a back pressure of 9.8 kPa were observed at 125 V applied voltage with 1 Hz frequency. Esashi et al. [1989] reported a three-layer piezoelectric pump with a flow rate of 15 μL/min and a back pressure of 6.4 kPa at applied 90 V with 30 Hz frequency. Olsson et al. [1995] reported a two chamber piezoelectric micropump to improve the performance. Koch et al. [1998] presented a piezoelectric micropump based on a screen printing of PZT (Lead Zirconate Titanate) on a Si membrane. A flow rate of 120 μL/min and a back pressure of 2 kPa were observed at applied 600 V with 200 Hz frequency. Schabmueller et al. [2002] fabricated a piezoelectric micropump with passive valves. A flow rate of 1500 μL/min and a back pressure of 1 kPa were achieved using ethanol. Feng and Kim [2005] reported a piezoelectric micropump that consisted of one-way parylene valves. A flow rate of 3.2 μL/min and a back pressure of 0.2 kPa were observed at applied 80 V with a lower power consumption of 3mW. Geipel et al. [2006] reported a novel design of micropump with a back flow pressure independent of the flow rate. The back pressure independence was reported up to 20 kPa at low frequency. Tree et al. [2009] reported a piezo stack actuated peristaltic micropump. A flow rate of 40 μL/min

was obtained at a frequency of 28.6 Hz using water. The flow rates were observed to be independent of back pressure up to 7 kPa, with a maximum back pressure of 45 kPa at 140 V. Wang et al. [2009] studied the effect of longitudinal flow asymmetry on pumping capability by using a simple pumping system comprising a piezoelectric buzzer embedded in a channel. Ali et al. [2010] studied the dynamic piezoelectric micropump process. The quantitative measurement of the pressure generated, applied electrical field, frequency and length of the actuator, were observed. Liu et al. [2010] proposed a disposable high performance piezoelectric micropump with four chambers in serial connection for a closed loop insulin therapy system. An outflow resolution of 6.23×10^{-5} mL/pulse was observed. A maximum back pressure of 22 kPa was reported at an applied voltage of 36 Vpp and 200 Hz frequency.

3. Specifications of DDS

The development of this work is mainly based on the specifications of an insulin delivery system. Table 1 shows the specification of a complete DDS. Only pump specifications are considered in this work.

Table-1 Specification of a complete DDS

Sl. No.	Type	Specification
1	Micropump	Valve less diffuser
2	Actuation	Piezoelectric
3	Supply voltage	<30V
4	Flowrate	< 10µl/sec
5	Needle length	<150µm
6	Needle inner diameter	10-20µm
7	Needle outer diameter	<100µm
8	Withstandable force	10N
9	Temperature	<50°C

4. Design

The conversion of mechanical energy to an electrical signal (voltage) and the reverse phenomenon is known as the piezoelectric effect. A stress applied to such materials will alter the separation between the positive and negative charges that leads to a net polarization at the surface. An electrical field with voltage potential is created in those materials due to the polarization. This property can be used to form the actuator, micropump, inkjet printer head, etc. Figures 3a & b show the top view and cross-sectional view, respectively, of the designed piezoelectric valveless diffuser micropump.

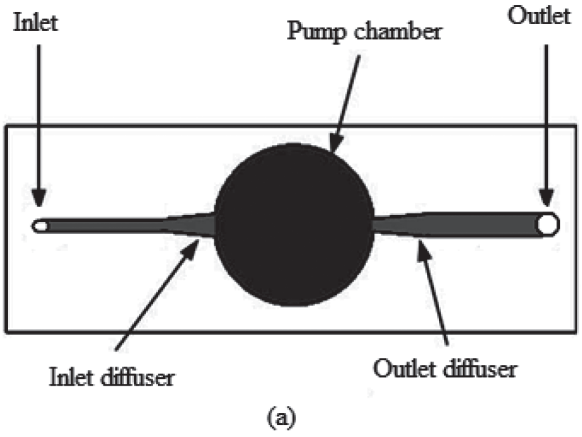


Fig. 3 (a) Top view of the piezoelectric valveless diffuser micropump

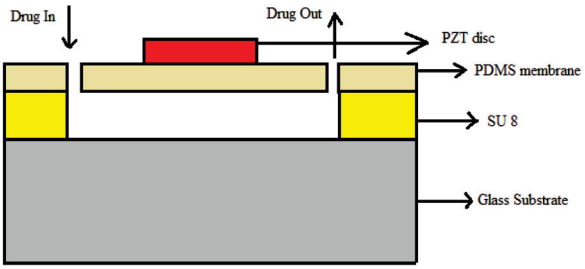


Fig. 3 (b) Cross-sectional view of the piezoelectric valveless diffuser micropump

The proposed system is composed of PDMS as membrane and SU-8 as pump chamber. The membrane of the micropump is actuated piezoelectrically. There are no check valves or active microvalves in the system. The flow of the medicine is controlled by the diffuser/nozzle element.

5. Numerical Analysis

For most membrane micropumps, the deflection of the membrane is reasonably small compared to the shortest characteristic length of the membrane. Therefore, the bending theory of thin plates is applicable and the governing equation for the deflection W, is

$$\left\{\frac{Eh^3}{12(1-\mu^2)}\right\}\nabla^4 w + h\rho_m\left(\frac{\partial^2 w}{\partial t^2}\right) = f_e - P \qquad (1)$$

Where E, ñ$_m$, h and ì are the elastic modulus, density, thickness and Poisson's ratio of the membrane, respectively, t is the time variable, P is the dynamic pressure exerted on the membrane by the fluid. P represents the coupling between membrane vibration and fluid flow. The flowrate of the device varies for the cases when $P \geq P_{out}$, $P \leq P_{in}$ and $P_{in} \leq P \leq P_{out}$ where Pin and Pout are the inlet and outlet pressure respectively.

6. Flowrate Calculation

Volume change can be expressed as

$$V = K_u V + K_p P \quad (2)$$

$$\Delta V = 2\pi \int_0^R w(r) r dr \quad (3)$$

Where K_p and K_u are only determined by the dimensions and material properties of the pump actuator. In actual pump applications, the applied voltage can be easily controlled while the pressure inside the pump chamber is coupled with the deformation of the pump actuator and the flow resistance of the whole system.

$w(r)$ = Total deflection due to applied voltage ($w_U(r)$) and pressure ($w_p(r)$).

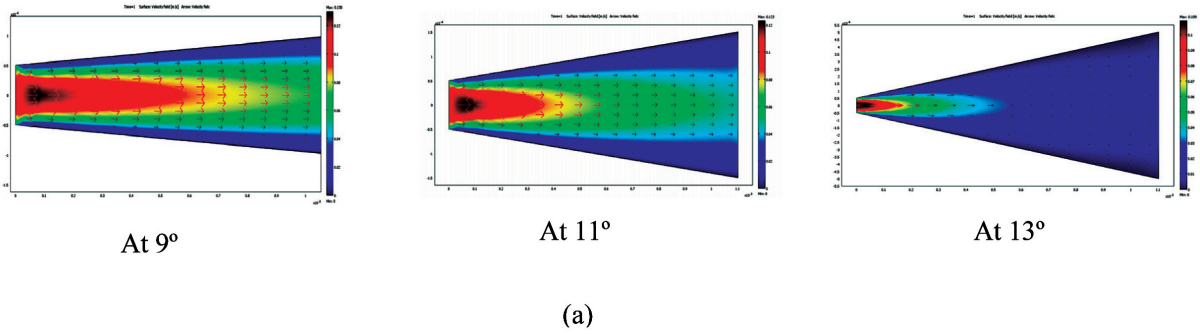
Maximum volume change = volume change due to total deflection

$$w(r) = w_U(r) + w_p(r) = \{^{w_1(r), 0 \leq r \leq a}_{w_2(r), a \leq r \leq b} + \{^{w_3(r), 0 \leq r \leq a.}_{w_4(r), a \leq r \leq b.} \quad (4)$$

So for the dimensions mentioned in the Fabrication section, the calculated net volume change = 9.22×10^{-12} m^3/ cycle. The micropump reported here is designed for a flowrate of 2µl/sec. So the operating frequency should be around 228 Hz.

7. Simulation Result

The diffuser of a valveless micropump plays an important role as the flow rate varies with the angle of the diffuser. It is observed from a simulation using FEM analysis-based Ansys that an increase in the diffuser angle increases the boundary layer which resists the flow. The simulation was carried out considering the diffuser angle as 9º, 11º and 13º. The 2D simulated results and the graphical representation of the simulation result are shown in figures 4a and 4b respectively. Red defines the maximum flow and blue is for the



(a)

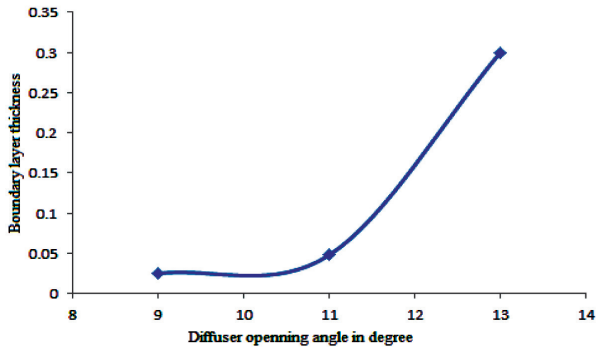


Fig. 4 (a) simulation results for the diffuser angle 9º, 11º and 13º, (b) diffuser angle Vs bondary layer

minimum. In this design, the diffuser angle was chosen as 9º.

8. Fabrication

The device dimensions are finalized through simulation to start fabrication of the device. Table 2 shows the device dimensions.

Table-2 The optimized device dimensions for the fabrication

Sl. No.	Parameters	Dimensions
1	PZT disc diameter	8mm
2	PDMS membrane diameter	9mm
3	PDMS membrane thickness	120µm
4	SU-8 chamber thickness	150µm
5	Diffuser length	1100µm
6	Diffuser inlet width	300µm
7	Diffuser outlet width	700µm

The fabrication of the PDMS membrane based SU-8 micropump has been carried out in three steps: (a) SU-8 pump chamber processing, (b) PDMS membrane processing, and (c) Bonding of SU-8 and PDMS.

(a) SU-8 pump chamber processing:

i. Spin coating: Initially ramping at 500 rpm for 10 sec and then ramped at 1000 rpm for 20 sec to achieve a thickness of ~125.

ii. The sample is then placed on an ultra-flat hotplate at 65ºC for 10 min and then at 95ºC for the next 30 min to complete the pre-exposure process.

iii. The SU-8 coated sample is then exposed using UV light through a mask plate for 24 sec.

iv. Next, the exposed sample is kept on the same hot plate for a post-exposure process at 65ºC for 1 min and then at 95ºC for the next 10 min.

v. The sample is then developed using an SU-8 developer for 6-7 min to realize the required structure. For proper developing, the sample is immersed in isopropyl alcohol. If the sample is under-developed, then white residue will remain on the sample, otherwise, clean structures will be visualized.

(b) PDMS Process:

a. Mixture of weigh 10 parts of Sylgard 184 pre-polymer and 1 part of curing agent.

b. Mix the above vigorously for a couple of minutes until the entire mixture is filled with bubbles.

c. The mixture is placed in a desiccator to degas (allow bubbles to rise out) for 30 minutes.

d. Pour the mixture slowly on the glass sample. Avoid bubbles and make sure that the glass sample is lying flat.

e. The coated sample is then spun at 600 to 800 rpm on a spinner.

f. The sample is then placed on a hot plate and cured at 80°C for 30 min.

g. The thin PDMS layer is then gently pulled off the glass substrate.

The SU-8 processed structure is shown in figure 5.

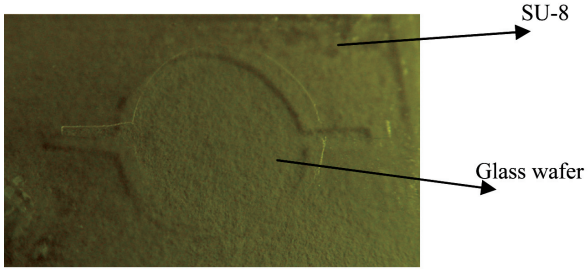


Fig. 5 Processed SU-8 on Glass wafer

(c) Bonding of SU-8 and PDMS:

The bond strength of PDMS/SU-8 is not very strong in a normal attachment. To improve the bonding between PDMS and the SU-8 structure, both of them are treated with oxygen plasma on a plasma sterilizer for 30 sec at 40 watt power. The PDMS layer is then gently placed on the SU-8 structure and kept in an oven at 80ºC for an hour.

The fabricated micropump is shown in figure 6 where the inlet and outlet are connected to the teflon tube through the teflon caps. These caps are attached using glue, MBond 200®.

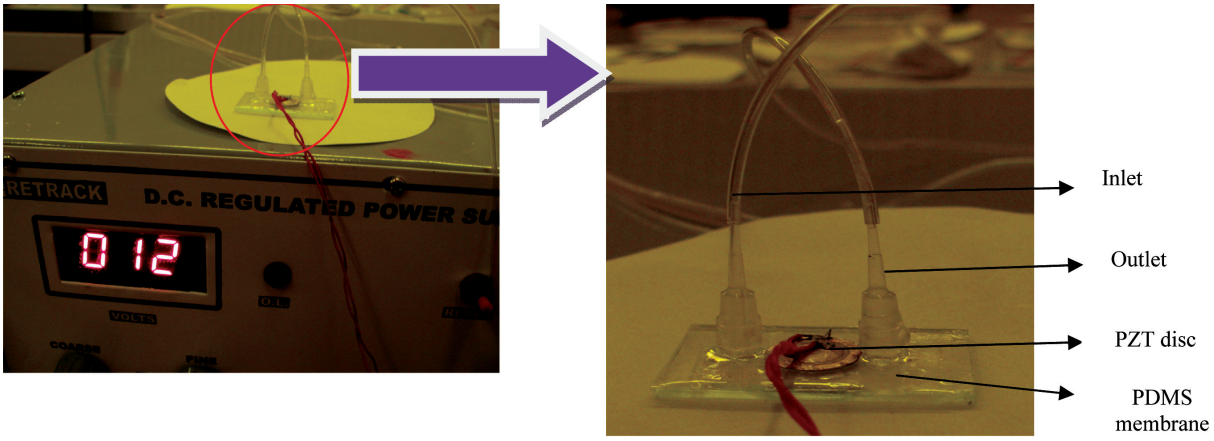


Fig. 6 The complete fabricated device

9. Testing and measurement

A commercially available PZT disc has been used for the actuation. The specifications of the PZT disc are shown in table 3.

Table-3 Electrical Specifications of PZT disc

Sl. No.	Properties	Specifications
1	Item No:	FT-12T-5.3B1
2	Resonant Frequency Fr	5.3 ±0.5KHz
3	Resonant Impedance R	500 W max
4	Capacitance C	13nf ±30% at 1KHz
5	Input Voltage Vp-p	30 max
6	Operating Temperature	-20 - +60ºC
7	Storage Temperature	-20 - +70ºC

The resonant frequency of the disc itself is 5.3 kHz. The PZT disc attached to a PDMS membrane is tested at dry conditions to achieve its maximum deflection at resonance frequency using MSA 400 of Polytec.

This disc-attached membrane uses pressure to push the fluid from the pump chamber to the outlet. At a particular frequency, the micropump gives the maximum fluid flowrate. It is also important to know about the peak at the resonance frequency of the Piezo disc-attached micro pump. Figure 7 shows the top and side view of the piezo disc and figure 8 shows the displacement magnitude at the resonance frequency. It is observed that the maximum displacement at the resonance frequency, 5009 Hz is 5.36 µm.

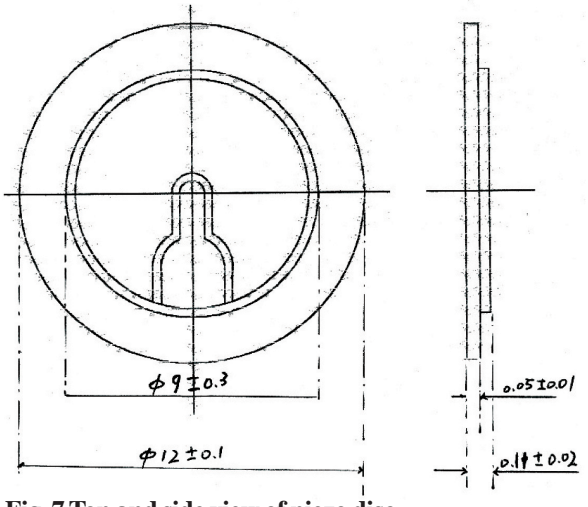


Fig. 7 Top and side view of piezo disc

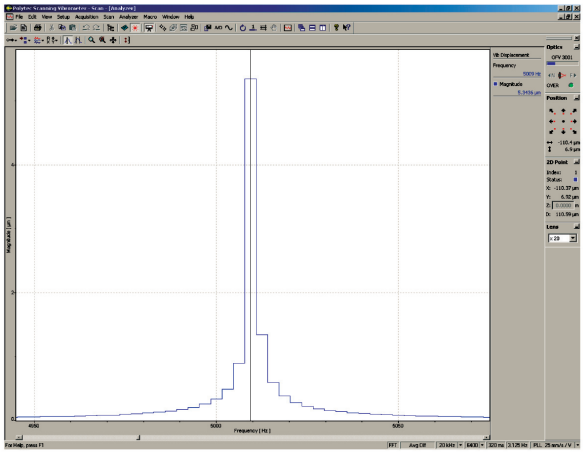


Fig. 8 The test result using MSA 400

The flowrate is measured from the water column displacement above the source water level. The multiplication of the tube length travelled by the water column in a unit of time and the tube's inner cross-sectional area gives the flowrate of the

micropump. The schematic view of the measurement setup is illustrated in figure 9. The figure shows that the inlet of the micropump is dipped into DI water. The voltage applied to the piezo actuator attached to the micropump raises the outlet water level above the container (inlet side) water level through the outlet tube.

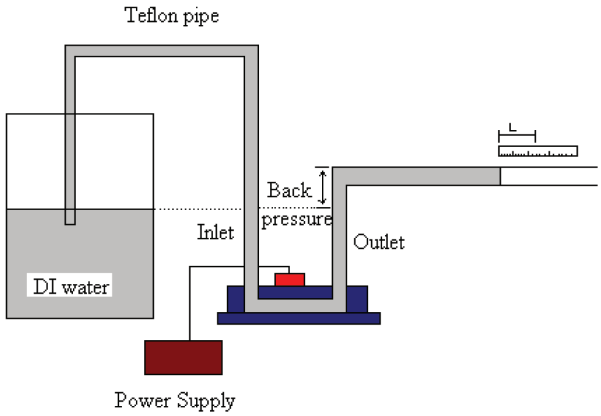


Fig. 9 Schematic view of the measurement setup

The flowrate of the micropump was characterized out using DI water and a maximum flowrate of 52 µl/min (or 0.867 µl/sec) was achieved at 30 V applied to the PZT disc at 5009 Hz frequency. The variation of the flowrate with applied voltage is shown in figure 10.

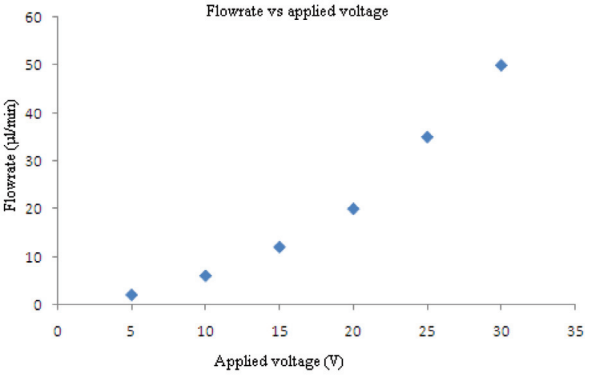


Fig. 10 The test result using MSA 400

10. Conclusions

The development of a drug delivery system is now the thrust area of Bio-MEMS research. A micropump is the major component of this system. Several research groups have developed different actuation-based micropumps. The micropump presented here is different from the others in three major aspects: (i) SU-8 is biocompatible, so no extra treatment is required to make the material biocompatible, (ii) PDMS membrane requires less force to deform and it is also biocompatible material, and (iii) no dry or wet bulk micromachining step is required. Frequency analyses have been carried out for this system. At the frequency of ~ 5009 Hz, the membrane deflection is greatest. The micropump is also tested using DI water and a flowrate of 52 µl/min (or 0.867 µl/sec) has been achieved applying 30 V to the PZT disc. The targeted flowrate was to be 2 µl/sec. This or a better flowrate can be achieved using a different PZT actuator. The micropump developed is going to be a remarkable component of a painless drug delivery system. Research is still going on to integrate the microneedle and the micropump to develop a painless integrated drug delivery system.

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References

Ali, M. Y., Kuang, C., An, J., Wang, G., 2010, A dynamic piezoelectric micropumping phenomenon, Microfluid. Nanofluid., 9: 385-396.

Ashraf, M. W., Tayyaba, S., Afzulpurkar, N., 2011, Micro Electromechanical Systems (MEMS) Based Microfluidic Devices for Biomedical Applications, Int. J. Mol. Sci., 12: 3648-3704.

Esashi, M., Shoji, S., Nakano, A.,1989, Normally close microvalve and micropump fabricated on a silicon wafer, In Proceedings of IEEE Micro Electro Mechanical Systems, An Investigation of Micro Structures, Sensors, Actuators Machines and Robots, Salt Lake City, UT, USA, 29-34.

Feng, G.H., Kim, E.S., 2005, Piezoelectrically actuated dome-shaped diaphragm micropump, J. Microelectromech. Syst., 14: 92-199.

Geipel, A., Doll, A., Goldschmidtboing, F., Jantscheff, P., Esser, N., Massing, U., Woias, P., 2006, Pressure independent micropump with piezoelectric valves

for low flow drug delivery systems, In Proceedings of 19th IEEE International Conference on Micro Electro Mechanical Systems, Istanbul, Turkey, 786-789.

Koch, M., Harris, N., Evans, A.G.R., White, N.M., Brunnschweiler, A., 1998, A novel micromachined pump based on thick film piezoelectric actuation," Sens. Actuators A: Physical, 70: 98-103.

Liu, G., Shen, C., Yang, Z., Cai, X., Zhang, H., 2010, A disposable piezoelectric micropump with high performance for closed-loop insulin therapy system, Sens. Actuators A: Physical, 163: 291-296.

Nisar, A., Afzulpurkar, N., Mahaisavariya, B., Tuantranont, A., 2008, MEMS-based micropumps in drug delivery and biomedical applications, Sens. Actuators B: Chemical 130: 917-942.

Olsson, A., Stemme, G., Stemme, E., 1995, A valve-less planar fluid pump with two pump chambers, Sens. Actuators A: Physical, 46-47: 549-556.

Schabmueller, C.G.J., Koch, M., Mokhtari, M.E., Evans, A.G.R., Brunnschweiler, A., Sehr, H., 2002, Self-aligning gas/liquid micropump, J. Micromech. Microeng., 12: 420-424.

Stemme, E., Stemme, G. , 1993, A valveless diffuser/nozzle based fluid pump," Sens. Actuators A: Physical, 39: 159-167.

Tree, F., Haeberle, S., Zengerle, R., 2009, Normally-closed peristaltic micropump with re-usable actuator and disposable fluidic chip, Procedia Chem., 1: 1515-1518.

Van Lintel, H.T.G., Van de Poi, F.C.M., Bouwstra, S., 1988, A piezoelectric micropump based on micromachining of silicon, Sens. Actuators A: Physical, 15: 153-168.

Wang, C., Leu, T., Sun, J., 2009, Asymmetrical flow effect applied to pumping performance of simple duct channel, Sens. Actuators A: Physical, 155: 203-209.

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