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Piezoelectric Micropump with Nanoliter Per Minute Flow for Drug Delivery Systems

(Pam Mikro Piezoelekrik dengan Aliran Nanoliter Per Minit untuk Sistem Penghantaran Bendalir Ubat)

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ABSTRACT

A Piezoelectric Actuated Valveless Micropump (PAVM) has been designed and successfully fabricated using MEMS fabrication processes. The micropump uses a PZT: Pb(ZrTi)O_x ceramic plate to actuate a silicon membrane which bends when a voltage is applied to the piezoelectric actuator. The resultant reciprocating movement of the pump membrane is then converted into pumping effect. By integrating dynamic passive valves into the device, the pump would then operate continuously with volumetric pumping rate determined by the frequency of the driving voltage. Simulation work to determine the micropump fluidic characteristics was performed using CoventorWare MemFSI™ module. The pump was fabricated on a double side polished silicon wafer via simple two-mask micromachining process. The fabricated micropump, having an outer dimension of 14 mm × 14 mm × 2 mm, was then tested with DI (deionized) water as the test liquid. A driving voltage of 16 V_{pp} was applied to the PZT actuator. Pump rate of 4.98 nL per min was obtained at 0.673 kHz. The fabricated micropump envisages a promising pumping method to be implemented into drug delivery systems.

Keywords: CoventorWare MemFSI; drug delivery; MEMS; PAVM; piezoelectric micropump

ABSTRAK

Pam mikro Janaan Piezoelektrik Tanpa Injap (PAVM) telah direka bentuk dan berjaya difabrikasi melalui proses fabrikasi MEMS. Pam mikro ini menggunakan plat seramik PZT: Pb(ZrTi)O_x untuk menggerakkan membran silikon yang membengkok bila voltan dikenakan pada penggerak piezoelektrik. Pergerakan salingan yang terhasil pada membran pam kemudiannya diterjemahkan kepada kesan pengepaman. Dengan mengintegrasikan injap-injap pasif dinamik kepada peranti, pam ini boleh beroperasi tanpa henti dengan kadar pengepaman ditentukan oleh frekuensi voltan pemacu. Analisis simulasi untuk menetukan ciri-ciri aliran pam mikro tersebut telah dilakukan dengan menggunakan modul MemFSI™ CoventorWare. Pam tersebut kemudiannya difabrikasi di atas wafer silikon dengan kedua-dua permukaan licin melalui proses pemesinan mikro dwi-topeng mudah. Pam mikro tersebut, yang bersaiz 14 mm × 14 mm × 2 mm, kemudiannya diuji menggunakan air ternyah ion sebagai cecair uji. Voltan pemacu 16 V_{pp} telah dikenakan pada penggerak PZT. Kadar pengepaman 4.98 nL per min telah dicapai pada 0.673 kHz. Pam mikro ini mencetus satu kaedah pengepaman yang berpotensi digunapakai dalam sistem penghantaran bendalir ubat.

Kata kunci: CoventorWare MemFSI; MEMS; pam mikro piezoelektrik; PAVM; penghantaran bendalir ubat

INTRODUCTION

The advent of modern photolithographic techniques and bulk micromachining has realized the miniaturization and mass production of today's semiconductor integrated circuit and microelectromechanical systems. Using BioMEMS devices, drug delivery into patient's body is improved since drug dosage could now be precisely delivered according to individual needs. The method also induces less pain and increases compliance to patient's body. Compared to the less precise traditional polymer-based controlled drug delivery systems, the integration of MEMS technology allows researchers to automatically deliver a wider range of drug mixture and precisely control the amount of drug released. Integration with integrated circuit hence allows programmable drug release. Batch fabrication techniques in MEMS enable the consistency and reproducibility of the device produced, which are important in pharmaceutical industry. Due to the compact size of a MEMS based drug delivery system, the drug supply is integrated onto the device. Moreover, because of the low fabrication cost, MEMS based systems are commonly disposable, hence ensuring hygienic application.

Microfluidics has been a rapidly developing field in MEMS since earlier this century (In-Stat MDR 2003). A survey conducted by In-Stat/MDR indicated that microfluidic devices comprise 35.7% of all MEMS devices produced in 2002, as shown in Figure 1 (In-Stat MDR 2003). MEMS-based microfluidics drug delivery devices in general include microneedles-based transdermal devices, micropump-based devices and biodegradable

MEMS devices. The miniaturization of the entire system, while often beneficial, is however not a requirement of a microfluidic system, since the term microfluidic refers to the micro quantity of the fluid which the device handles.

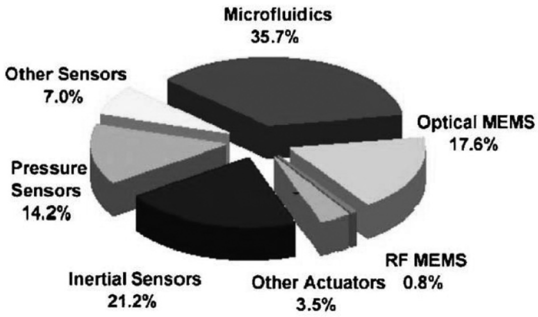


FIGURE 1. Breakdown of MEMS devices categories (In-Stat MDR 2003)

Micropump-based devices are perhaps the most extensively researched topic in microfluidics system, which covers the fundamental role in drug delivery. The micropump precisely controls the dosage of drug to be delivered into the body. For instance, a small and compact pump can be used to administer drug delivery for the treatment of diseases like diabetes, in which a constant level of insulin needs to be maintained in the body. The micropumps can be programmed to administer insulin at a constant rate throughout the day, thus eliminating any surges or deficits of the drug in the patient's bloodstream (Amirouche et al. 2009; Nisar et al. 2008). Modern pumps utilize microfabrication techniques to fabricate them at miniature sizes with precise specifications (Van Lintel et al. 1988).

Van de Pol and Van Lintel (1990) presented the first silicon micropump based on thermopneumatic actuation of a thin membrane. Many micropumps have since been developed in the last few years based on different actuating principles and fabrication technologies (Nguyen & Wereley 2002). However, pump output characteristics and other functionality factors are yet to be investigated thoroughly.

Mechanical reciprocating displacement micropumps with vibrating membrane have generated the most interest. Even though various actuating principles of membrane micropumps such as piezoelectric, pneumatic, thermal, electrostatic, and electromagnetic have been developed, the piezoelectric type has been widely used (Laser & Santiago 2004). Piezoelectric Actuator Valveless Micropump (PAVM) is deemed suitable for use in controlled drug delivery application (Jeong et al. 2005). The pumps are widely used due to their ability to conduct particles in the absence of interior moving mechanical parts (i.e. active valves), thus having less risk of clogging (Woias 2001; Woias 2005). Besides, dynamic passive valves, or commonly known as nozzle-diffuser valves, are preferred due to its durability upon prolonged usage and ease of manufacture using conventional silicon micromachining techniques.

This paper presents a PZT-based micropump which operates via reciprocal motion of the PZT bonded membrane. The mechanical properties of the voltage driven membrane was analyzed using CoventorWare MemFSI™ simulator. The micropump is then fabricated using standard MEMS fabrication processes.

DESIGN OF MICROPUMP

The PZT-based micropump designed is a fluid-structure coupled system consisting of a piezoelectric actuator, silicon membrane, pump chamber and passive valves, as depicted in Figure 2.

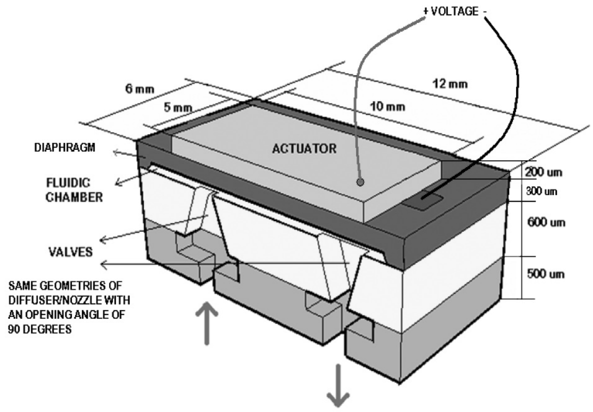


FIGURE 2. Cross sectional view of the designed PAVM

The piezoelectric micropump is of the reciprocating type, in which the pumping effect comes from the combination of the reciprocating movements of the membrane and the design of the valves. The pump uses diffuser and nozzle pair instead of check valves for flow rectification. When an electric field is applied to the PZT plate, the strain produced in the PZT plate causes the passive plate (i.e. silicon membrane) to expand or contract, resulting in actuation of the micropump. The actuator drives the membrane to increase or decrease the volume of the pump chamber periodically. Piezoelectric actuator vibration produces an oscillating flow and alters the chamber volume via curvature change of the membrane. The deflection is transformed into pumping effect that drives the fluid inside the pump chamber through the inlet and outlet.

The nozzle and diffuser elements, which are a type of as dynamic passive valves, act as the inlet and outlet respectively. In a reciprocating micropump, the nozzle and diffuser pair acts as a diffuser during half of the membrane actuation cycle and as a nozzle during the other half of the cycle (Stemme & Stemme 1993). During pumping, the directional effect governs the net pumping volume for each oscillating cycle, since the nozzle and diffuser pair is in opposite orientation with respect to each other.

The pump design presented in the study uses diffuser/nozzle structures with larger opening angle of approximately 54.7°. Since this opening angle is large, the nozzle is the positive direction of flow and the diffuser is the negative direction of flow (Figure 3). This is due to the fact that a diffuser/nozzle element is geometrically designed to have a lower pressure loss in the nozzle direction than in the diffuser direction for the same flow velocity.

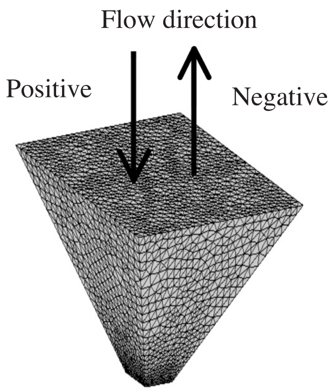


FIGURE 3. Definition of the flow directions

During the increase of the chamber volume (supply mode), the inlet element act as the nozzle, with a lower flow restriction than the outlet element, which acts as a diffuser. This means that a larger volume is transported through the inlet into the chamber than through the output. During the decrease of chamber volume (pump mode), the outlet element acts as a nozzle with a lower flow restriction than the inlet element, which acts as a diffuser. This resulted in a larger volume transported through the outlet of the chamber than through the inlet, as shown in Figure 4. The resulting volume of a complete pump cycle is the net volume is transported (pumped) from the inlet side to the outlet side, despite the fact that the diffuser/nozzle elements convey fluid in both the directions.

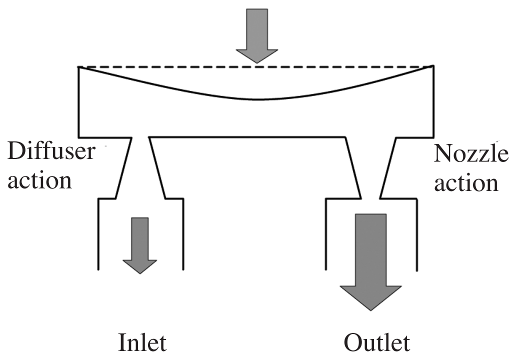


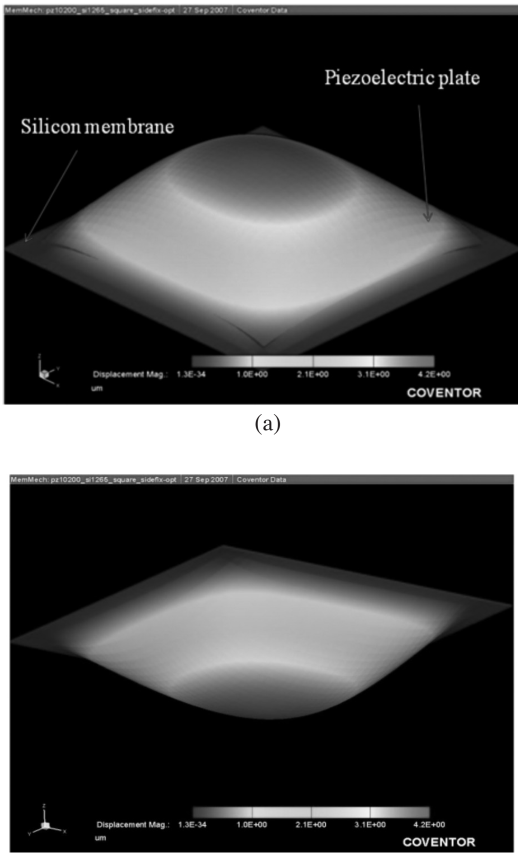
FIGURE 4. Operating cycle of the dynamic passive valves (diffuser/nozzles) pump during pump mode

SIMULATION OF MICROPUMP CHARACTERISTICS

In order to accurately predict the working behavior of the micropump, the fluid-structure dynamic coupling effects were taken into account. Based on the membrane vibration equation and Navier Stokes flow equation, accurate fluid-structure interaction (FSI) involving electro mechanics, solid mechanics and fluid mechanics of the system is analyzed. The piezoelectric domain, the structural domain and the fluid domain are coupled together in the simulation. The fluid structure interaction is the governing principle of the displacement pumping. This method is applicable since the structure which surrounds the fluid is flexible in response to the motion of the fluid.

Numerical handling of the coupled problems is the major challenge in numerical analysis step. Since the full 3D numerical solution is extremely complex, it is ideal to perform 3D-FSI simulations in order to simulate the micropump in details. CoventorWare MemFSI™ module is used to simulate the pumping behaviors.

The simulation results shown in Figure 5a and Figure 5b are the preliminary stages in determining the optimized dimension for the silicon membrane as the piezoelectric dimension is fixed at 10 mm × 10 mm × 200 μm. The optimized dimension obtained for the silicon



(a)

(b)

FIGURE 5. (a) Exaggerated view of +4.3 μm membrane displacement at a driving voltage of +100 V and (b) Exaggerated view of -4.3 μm membrane displacement at a driving voltage of -100 V

membrane is 12 mm × 12 mm × 100 μm. All four edges of the silicon membrane are fixed in all directions and the bonding effect between the two layers is ignored. A voltage potential is applied on the actuator while the silicon membrane is grounded.

The simulation results indicate that the displacement profile of membrane is a parabolic surface, in which the maximum value occurred at the center of the membrane, and the minimum value occurred along the fixed boundary.

FABRICATION PROCESS

The piezoelectric micropump was fabricated in IMEN-UKM. In the process, <100> oriented silicon has been used as the substrates for both the membrane and the valves. The substates are 650 μm thick, double side coated with 200 nm silicon nitride layers. For both first and second substrate, double side silicon etching was performed using two masks. Prior to the etching, the nitride layers on both sides of the substrate are patterned using DRIE. The patterns are used as the masks for the KOH etching. KOH solution (35%) is prepared by mixing a 42.76g potassium hydroxide (KOH) pellets into a 100 ml H_2O. The solution is heated to 80°C, at which the simultaneous double side etching process is performed. The silicon substrates were anisotropically etched on both sides to create the pump membrane and the nozzle/diffuser valves. Double side alignment is utilized since both the inlet and outlet valves and the pump chamber are simultaneously etched. To complete the micropump structure, both completed substrates are bonded together using epoxy glue, as depicted in Figure 6.

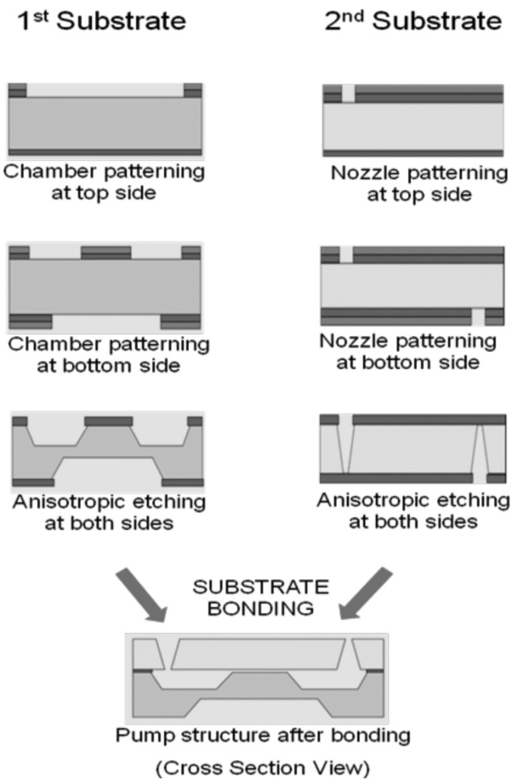


FIGURE 6. Schematic process steps for fabricating micropump using bulk micromachining

The etched membrane is depicted in Figure 7. A smooth and clean 275 μm silicon trench is produced at an angle of 54.7° from horizontal. The KOH solution (35%) at 80°C produces a controllable silicon etch-rate of approximately 1 μm per minute. Using this rate, a 100 μm thick membrane would require 275 μm etch on each side of the substrate. The similar process sequence is used to create the nozzle/diffuser valves on a different substrate. 650 minutes of KOH etching is required to completely etch the silicon for creating the valves. Figure 8 shows the completed micropump with membrane and valves bonded together.

RESULTS AND DISCUSSION

CoventorWare simulation enabled the effective fluid flow rate to be determined for different membrane actuation voltages using the optimized design parameters tabulated

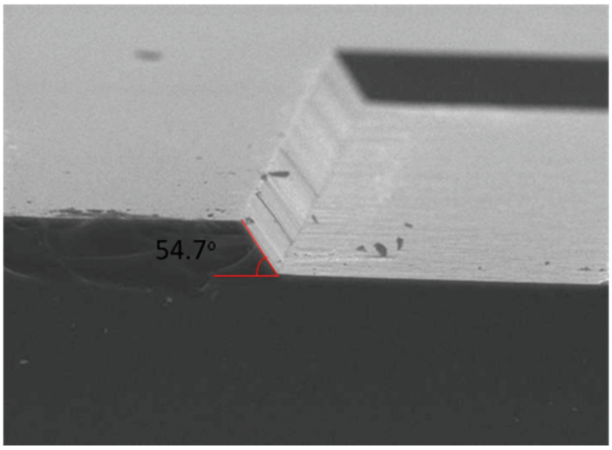


FIGURE 7. The fabricated silicon membrane. KOH etching wall produces 54.7° angle from horizontal

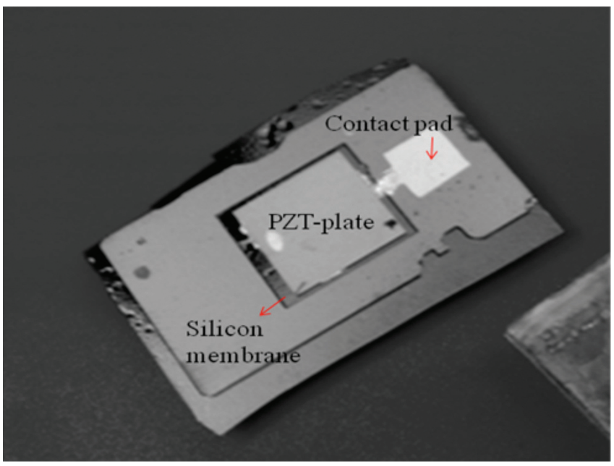


FIGURE 8. Fabricated piezoelectric micropump

in Table 1. The design parameters are determined based on fabrication capabilities and simulation results. The FSI simulations show that during pumping, the dominant flow is at the outlet while during suction, the dominant flow is at the inlet, as depicted in Figure 9.

TABLE 1. Parameters of the designed micropump

Elements	Value
Membrane width	6 mm
Membrane thickness	100 μm
Silicon Young's Modulus	169 GPa
Silicon Poisson's ratio	0.3
Actuator width	5 mm
Actuator thickness	200 μm
PZT Young's Modulus	55.55 GPa
PZT Poisson's ratio	0.3
Pump chamber depth	100 μm
Pump chamber width	6 mm
Valve height	650 μm
Valve min width	200 μm
Valve max width	1156 μm

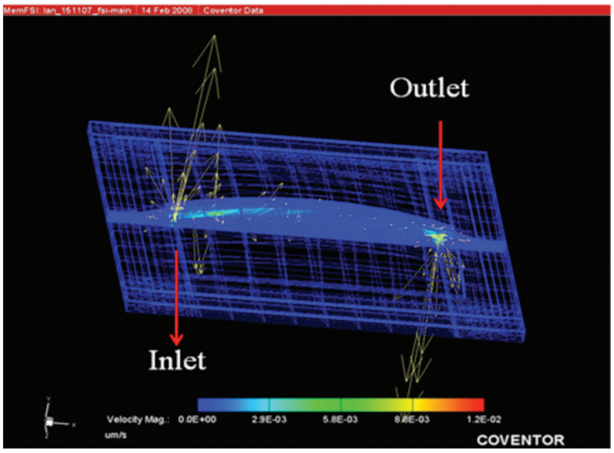
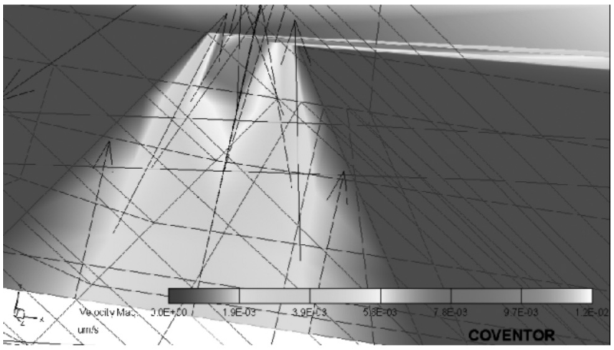


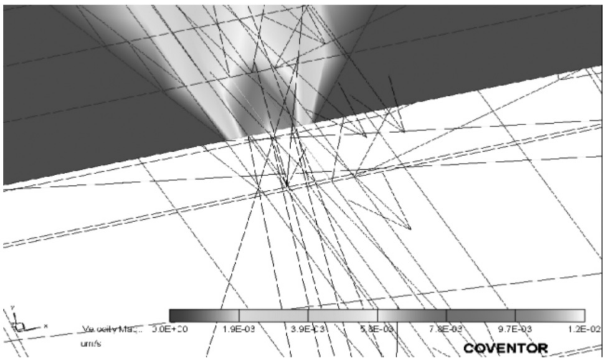
FIGURE 9. MemFSI™ simulation showing velocity vector of fluid flow

Simulation results show that deflection achieved is sufficiently large to enable the fluid to flow through the pump chamber. The net fluid flow direction is towards the outlet, and the velocity vector confirmed the flow type movement is laminar (Figure 10). Simulation done in this work on a piezoelectric micropump can be easily adapted into various geometries and pump configurations, allowing researcher to analyze the behavior of the flow in a wide range of microfluidic devices.

The simulation results indicate that the maximum displacement and velocity of the membrane occurred at the central region, while the minimum values occurred along



(a)



(b)

FIGURE 10. Velocity vector of fluid flows in the micropump. Note the higher pressure at (a) the inlet during the supply mode and (b) the outlet during pump mode

the boundary. The displacement behavior is consistent with analysis performed on the square membranes of MEMS polysilicon encapsulation and MEMS microphone in previous work (Hamzah et al. 2004; Hamzah et al. 2006). The magnitude of vibration of the membrane is determined by the driving voltage and the thickness of the PZT plate. Further analysis reveals that the displacement of the membrane would increase with driving voltage and decrease with PZT thickness.

Figure 11 shows that the membrane displacement increases proportionally to the driving voltage. On the other hand, the maximum displacement of the membrane is determined by the membrane thickness. This phenomenon is due to fact that the membrane is becoming exponentially easy to deflect with decreasing thickness. The mechanical properties of silicon shift from rigid to flexible with decreasing thickness, as depicted in Figure 12.

The fabricated piezoelectric micropump was tested using DI water as the fluid medium. Fluid flow measurement test was set-up to measure the fluid flow rate (Figure 13). Two of the most important parameters that affect flow rate, namely driving voltage and driving frequency of the micropump were examined. It is observed that by increasing the driving frequency, the flow rate can be increased. The driving frequency however must not exceed the resonant frequency of the membrane. In this study, at a driving voltage of 16 Vpp and optimal frequency

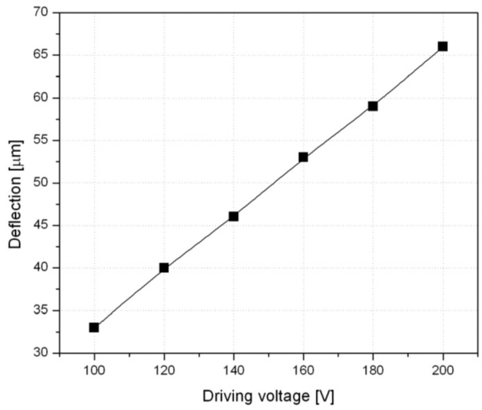


FIGURE 11. Correlation between membrane deflection and driving voltage

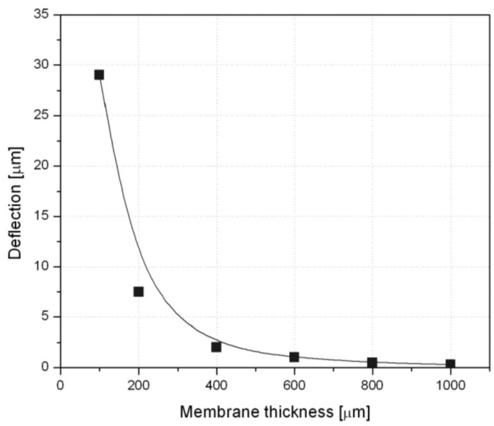


FIGURE 12. Correlation between membrane thickness and deflection

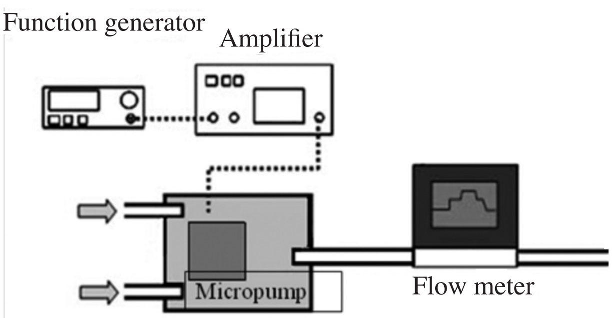


FIGURE 13. Fluid flow measurement test set-up

of 0.673 kHz, the optimal flow rate is 4.98 nL per min. This result indicates that an optimal working frequency can be obtained, at which the flow rate of the micropump achieves the maximum when the external electric voltage is fixed. The experimental findings envisage the potential usability of this micropump in microfluidics and drug

CONCLUSION

A piezoelectric micropump with nozzle/diffuser elements was fabricated using simple MEMS fabrication technologies. The micropump was designed to deliver fluid at a precisely controlled rate. Double sided etch technique was used in the fabrication process. By combining KOH etched silicon membrane and dynamic passive valves, a dynamic micropump was successfully fabricated. The magnitude of membrane deflection is determined by the driving voltage and the thickness of the PZT plate. The fabricated micropump ideally operates at 0.673 kHz, producing a net flow rate of 4.98 nL per min. The work presented here illustrates the feasibility and merit of utilizing simple MEMS processes for fast and reliable fabrication of valveless micropumps. The fabricated micropump envisages potential usability and integration with microfluidics and drug delivery systems.

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