A Low Cost Polymer Based Piezo-actuated Micropump For Drug Delivery System

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ABSTRACT

This paper presents the design, realization and simulation of a novel polymer based check-valve micropump actuated by piezoelectric disc. Comparing with silicon substrate, polymer materials have such advantages as flexibility, chemical and biological compatibility, 3D fabrication possibility and low cost in materials and mass production. Laser micromachining technology and precision engineering techniques were used to fabricate the prototype with the dimension of Φ18mmx5.2mm. Results of preliminary experiments on fusion bonding between polyimide and polycarbonate are also presented. Using DI water as the pumping medium, the presented micropump is expected to achieve self-priming, bubble tolerance and low power consumption and a flow rate of 30μl/min at the resonance frequency of 300Hz.

Keywords: polymer, laser microfabrication, fusion bonding, microvalve, flow rate.

1 INTORDUCTION

Interest in polymer materials has been motivated recently in the MEMS device for real world application purpose. Polymer materials, such as polycarbonate (PC), polyimide (PI), PMMA and polyester (PE) etc, have the potential abilities to be MEMS materials due to their properties such as flexibility, chemical and biological compatibility, 3D fabrication possibility and low cost in materials and mass production [1].

A micropump as one of the devices in the microfluidic system for bio-medical application is able to supply steady controllable fluid flow at the range from micro liter to milli liter per minute. There is a greatly growing market for the treatment of some diseases which need the micro dosing system [2].

This paper presents a low cost prototype of a polymer based self-priming micropump for drug delivery application. The micropump is actuated by the piezo-electrical disk while the pump chamber and the check-valves were machined in polycarbonate (PC) and polyimide respectively. The proposed micropump has the potentials to be fabricated by hot embossing technology for mass production purpose, which will reduce the fabrication cost. The actuation unit is driven by the controller unit which consists of an AC power supply and a function generator.

2 PUMP DESIGN

The micropump consists of a piezoelectric actuator, polycarbonate pump housing and two polyimide check valves. The working principle is shown in Figure 1[3].

When the actuator diaphragm is relaxed, both of the check valves are closed and there is no flow in the chamber. During the supply mode, the piezo disc is excited and the diaphragm is protruded caused by the deflection of the piezo disc. Thus, an under pressure is generated in the chamber which consequently generates a pressure difference across the inlet valve and the out valve. This pressure difference makes the inlet valve open and the outlet valve close

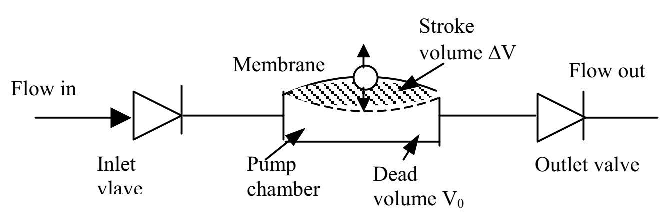


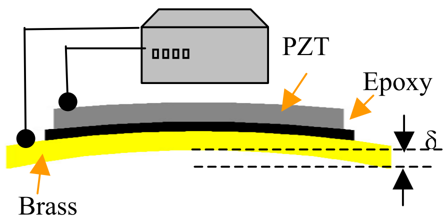
Figure1 The working principle of the check valve pump.

Hence, the flow (gas or liquid) comes into the pump chamber till the inlet valve closed by the inversed pressure difference generated by the actuator diaphragm. Then, the supply mode ends and the pumping mode starts. As the actuator diaphragm is depressed with the deflection of the piezo disc, an overpressure will be generated in the chamber. The flow goes out from the chamber through the opened outlet valve.

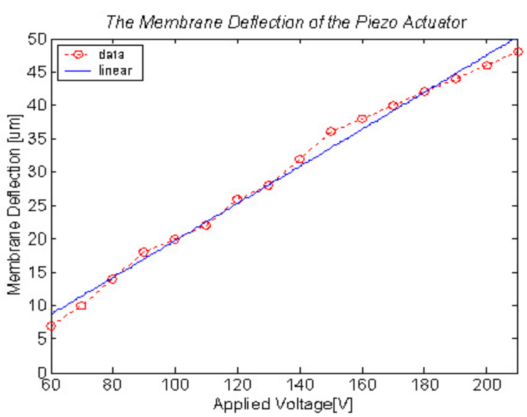
3 THE ACTUATOR UNIT

The actuator was made of a piezoelectric disc with the dimension of ϕ16mm × 0.2mm thick and a brass membrane with the dimension of ϕ16mm × 0.1mm thick. The piezo ceramic disc and the brass membrane were joined together by the Loctite epoxy adhesive (E-90FL).

When the piezo of the actuator is applied with an electric field of U/t, where U denotes the applied voltage and t the piezo disc thickness, the movement along the bonded surface of the piezo will be prohibited. Hence, it will generate forces and moments to lead to the curling of the brass membrane shown in Figure. 2(a). A longitudinal deflection larger than the horizontal change along the length of the piezo disc is obtained.



(a) Excitation schematics



(b) The deflection variation with applied voltages

Figure.2. The piezo actuator working performance

From the plate and shell theory [4], the displacement *δ(r)* at any radial point r can be arrived at:

$$\delta_r = \lambda \frac{d_{31}(R^2 - r^2)}{2t^2} U \qquad (1)$$

where R denotes the radius of the actuator. The mechanical loading, the elastic properties of the brass membrane are assumed to be neglected for Equation (1). The constant λ is achieved empirically. From Equation (1), the maximum deflection is achievable at the centre of the disc and the deflection variation is linear with the applied voltage. The testing result of the maximum deflection variations of the actuator diaphragm with applied was shown in Figure. 2(b). A linearity of the variation between the maximum deflection and the applied voltage (larger than 60V) was observed through a dial indicator with 2μm accuracy.

Integrating Equation (1), we can arrive the pump flow rate at:

$$Q = \lambda \cdot \frac{\pi}{4} \cdot \frac{d_{31} R^4}{t^2} U \cdot f \qquad (2)$$

where, f is the frequency of the excitation voltage. Therefore, the flow rate can be linearly controlled through changing the applied voltage and the excitation frequency. It offers the fundamental of the pumping working principle.

4 REALIZATION

The valve foil has a thickness of 50μm and it is fabricated using ultra-short pulse femtosecond laser micromachining technology. Femtosecond laser micromachining is a rapidly advancing area of ultra-short laser applications. It utilizes the ultra-short laser pulse properties to achieve an unprecedented degree of control in sculpting the desired microstructures internal to the materials without collateral damage to the surroundings. Because typical heat diffusion time is in the order of nanosecond to microsecond time scale whereas the electron-phonon coupling time of most materials are in the picosecond to nanosecond.

The reason for selecting polyimide as the material for the valve is because it has a relatively low Young's modulus (steel and silicon: 210GPa, polyimide: 30GPa). Hence, the critical pressure required to switch the valve flap is small and a larger gap (and flow rate) can be obtained. Therefore, it is not difficult to satisfy the basic operating conditions for the operation of the pump is easy to achieve. Also, polyimide has high resistance to corrosion. In addition, polyimide has good adherence with metals and has good sealing effect. All these mechanical and chemical properties make it as a suitable material.

For this micro-pump structure, the dead volume of the pump chamber comprises of the inlet and outlet and the small part encapsulating the inlet valve. A unique and distinctive feature of this micro-pump is the absence of a chamber immediately beneath the actuating diaphragm. This design yields a much smaller dead volume. Therefore, the compression ratio of the pump, which is the ratio of the stroke volume of the actuator unit and the dead volume will be improved. The higher compression ratio will help to achieve self-priming and bubble tolerance.

Polymer fusion bond technology was applied to assemble the pump structure. The bonding process was finished in a clean room oven with the applied pressure of 282.7 KPa, the temperature of 125°C for 80 minutes. Due to the small difference between their glass transition temperatures (T_g) [5], the polyimide and polycarbonate could be bond together relatively easily. A special fixture for the bonding process has been designed and made to control the lateral and vertical expansion less than 1.5%.

After that, the samples were tested with an Instron™ tensile testing equipment. The experiment setup is shown in Figure 3. Two roughened steel bars were fixed by the epoxy at the both sides of the sample as the holders for the equipment. It was observed that the fracture always occurred at the interface with epoxy. It means the bond strength is higher than the epoxy strength. The bonding interface was scanned by the Scanning Acoustic Microscopy (SAM). A bubble-free uniform bond interface is achieved. The bonding mechanism at the interface is still under study by the software of Cerius2®.

A piezoelectric actuator is laminated to the top of the pump housing to form a chamber. The check valves are fabricated in the polyimide membrane and bonded to the pump housing. The polyimide material has high glass transition temperature and flexibility, strong high anti corrosive coefficient, and high elasticity. Polycarbonate (thermo set) has been widely designed and used in the domestic and commercial market due to its good corrosion resistance and biocompatibility. And, using the polycarbonate as the pump housing material can improve the rigidity to increase the life time of the pump housing.

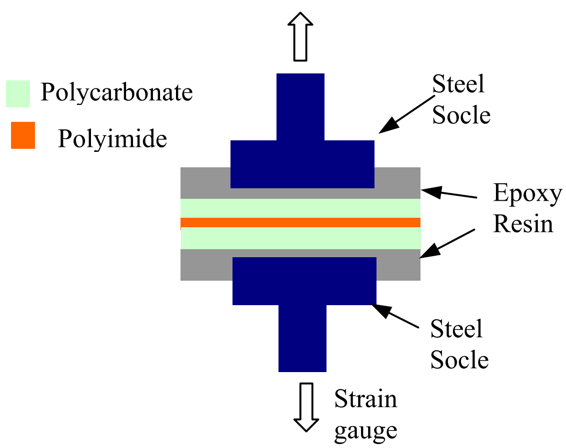


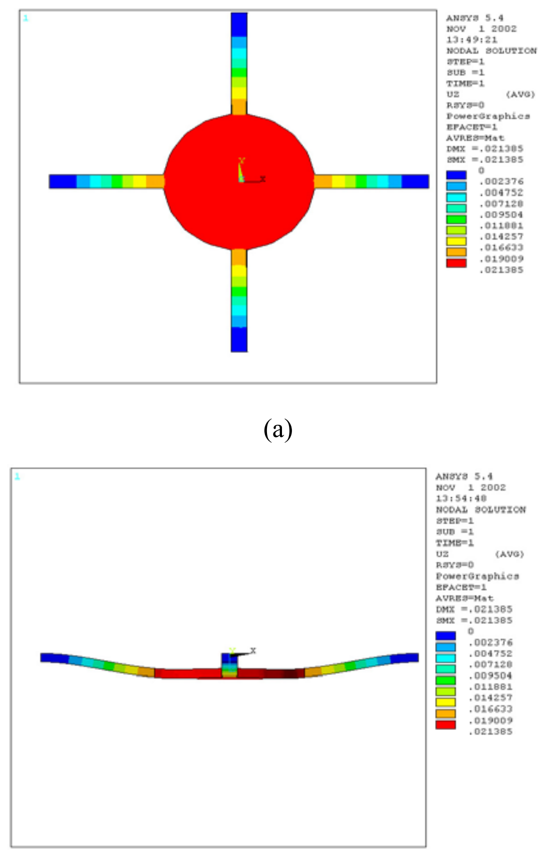
Figure 3 The experiment setup for the tensile strength test.

The pump structure, inlet and outlet were fabricated in polycarbonate by conventional precision engineering. The sample was successfully completed with the dimension of Φ18mmx5.2mm. A syringe was used to achieve a bubble-free chamber prior to the test. De-ionized water was used as the pump medium.

5 MODELING OF THE VALVE UNIT

A finite element analysis of the valve deflection was carried out using ANSYS™ to study the basic knowledge of the stress distribution and the static working condition. Some assumptions were made to simplify the model. First, the fluid pressure was considered uniformly loading on the valve unit. And, it was modeled as a simply supported beam with the boundaries at the end of the beams. The following parameters were used for the analysis:

FEA Property:	3D
Analysis Method:	Static
Pressure Load Area:	Uniformity on Membrane
Material:	Polyimide
Young's Modulus :	2400MPa
Poisson Ratio:	0.35
Density:	1420 kg/m^3
Pressure:	10^4 Pa



(a)

(b)

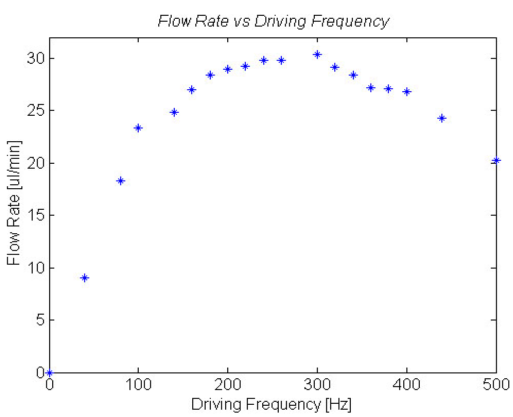
Figure 4. The modeling result of the valve deflection, (a) top view, (b) side view

The deflection result of the valve unit is shown in Figure 4. It was observed that the maximum deflection at the center point is about 21.4 μm under the pressure of 10^4 Pa. As the chamber height is about 500μm, for the safety consideration, the valve unit will not collapse with the actuators during the supply mode.

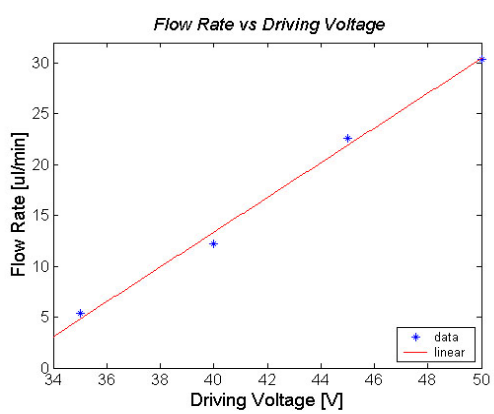
6 TESTING RESULTS

The sample was successfully completed with the dimension of Φ18mmx5.2mm. A syringe was used to achieve a bubble-free chamber prior to the test. De-ionized water was used as the pump medium.

When the pump was applied with a square pulse voltage (50V) at the pump head of (H=20mm shown in Figure 5). The flow rate rises almost linearly with the increase of the driving frequency till it reaches the maximum value of about 30.35μl/min. The resonance frequency was observed at 300Hz (Figure 5a). A wide working range from 200Hz to 450Hz was obtained to offer a flow rate of 25-30ul/min. The flow rate also increases very linearly with the variation of applied voltage in the range of less than 50v (yield voltage of piezo disc) (Figure 5b). A pumping head was achieved at 350mm with 150V and the power consumption of this pump was 15mW working at the 50V.



(a)



(b)

Figure 5. The testing result of the pump flow rate, (a) Flow rate variation with frequency, (b) Flow rate variation with driving voltage

7 CONCLUSION

The design, fabrication and testing of a polymer-based piezo-actuated micropump has been illustrated, which has the feasibility in the application of drug delivery system. Fusion bonding technology was tested successfully to bond the polymer materials at the appropriate temperature and pressure.

Using DI water as the pump medium, the maximum flow rate of 30.35μl/min has been achieved at the resonance frequency of 300Hz and the applied voltage only 50V. The pumping head can achieve 350mm and the power consumption is only about 15mW at 50V. The micropump's life time is more than 120hrs.

The micropump model would be developed further and the empirical constants determined at a later stage. The low production cost makes this micropump great potential to the disposable usage in the drug delivery systems.

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