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A PDMS Diaphragm Micropump Using Electroosmotic Actuation

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Abstract

This paper presents a novel micropump using two EO (electroosmotic) actuators toward biomedical systems and drug delivery systems. This micropump consists of two EO actuators, one indirect pumping mechanism, and two check valves. The indirect pumping mechanism includes two sealed chambers, one is the driving fluid chamber and the other is working fluid chamber. The EO actuators with the impulse input make the polydimethylsiloxane (PDMS) diaphragm which separates the two chambers reciprocate. So the working liquid can be pumped from inlet to outlet. Due to the large deflections of the PDMS diaphragm, the device is completely self-filling and able to pump liquid even operated in vertical situation. By using the indirect pumping mechanisms, the EO actuator can be used more widely without considering the properties of the working liquid. The operation of the micropump was investigated experimentally, as well as through simulations using CFD (commercial fluid dynamic) analysis programs. And a maximum flow rate of 133μl/min (using top water as working liquid) was obtained by an input voltage of 30V and an exciting frequency of 1.4Hz. Superior pumping performance was achieved when the duty ratio of the voltage applied to the EO actuators was 30%. And some experiments were manipulated to show the characteristics of the PDMS diaphragm and the check valve. These experiments results demonstrate the diaphragm micropump performance and ability to deliver the fluid and drugs to biomedical systems.

Key Words: Electroosmotic actuation, Microfluidic system, Drug delivery system, Diaphragm micropump.

1. Introduction

During the last two decades, with rapid development of micro-electro-mechanical systems (MEMS) and MEMS technologies having been applied to the needs of biomedical industry, a new emerging field called microfluidics has been brought about [1]. The development of microfluidic systems have been widely used in variety of fields. Microfluidic systems are mainly used for chemical analysis, biological and chemical sensing, drug delivery, molecular separation such as DNA analysis, amplification, sequencing or synthesis of nucleic acids and for environmental monitoring [1]. Taking drug delivery for example, most of the drugs are effective if they are delivered within a specific range of concentration between the maximum and minimum desired levels. Above the maximum they are toxic and below that range they have no therapeutic benefit [2]. So a microfluidic system can be used to precisely control flow rates and calculate effective amount of drug.

As a very important microfluidic element, various types of diaphragm micropumps are developed since the demand of microfluidic system. As we know, various kinds of diaphragm materials and actuation methods have been used for micropump to control the flow rates of the working liquid [3]-

[7]. But no pump is perfect. Each pump has their particular advantages and disadvantages [8]. The diaphragm micropump exploits the periodic volume strokes of the actuator diaphragm by using the existing actuation methods such as piezoelectricity, thermopneumatic, electrostatic, and magnetic and so on.

Van Lintel et al. reported a first attempt to fabricate silicon micropump based on piezoelectric actuation [9]. Since the first PZT diaphragm micropump was successfully fabricated, a lot of researches have been done for the PZT actuation method. Although they have high performance, the major limitation of the PZT actuated micropumps is the requirement of high voltages. In addition, the micropump actuated by PZT is difficult to be produced in blocks for its complex structure [15]-[17].

Van De Pol et al. reported a thermo pneumatic which takes advantage of the pressure of the air chamber changing caused by the heater to induce a deflection of the diaphragm and pump out the liquid [10]. Its advantage is low input voltage. However, because of heat transmission, the temperate rise of the drugs which are delivered by the thermopneumatic pump may happen so that the characteristics of the drugs may change. In addition, the performance of the thermopneumatic pump was highly affected by ambient temperature [17]-[22].

In recent years, PDMS diaphragms have been widely used in microfluidic systems for the flow actuator and the flap valve because of its advantages such as a transparency, a remarkable mechanical behavior, simple structure bonding process and a low production cost [11]-[13]. So PDMS diaphragm is the optimal option to fabricate the diaphragm micropump.

In our work, a PDMS diaphragm micropump based on EO actuators was designed and fabricated. As shown in Fig.1, a bed-packed EO actuator (manufactured by Nano Fusion Technology Inc) was used as force generator for the advantages of low input voltage (0-30V) and high output pressure (0-200kPa). However, the major limitation of this kind of EO pump is that outlet characteristics heavily depends on the properties of the pumped fluid that flow through the EO actuator. The combination of the EO actuation approach and a PDMS diaphragm could solve the above mentioned problems. So an improved diaphragm micropump based on two EO actuators and an indirect mechanism with PDMS diaphragm was designed with the goal of low input voltage, portability and widely using for different kinds of drug delivery system. In addition, by using passive check valves, backward flow and fluid leakage were blocked. Unlike planar micropump, this micropump can be used with any posture such as horizontality, verticality and other states. It is more fitting for different working environment. The experimental results show that the system is very useful to deliver the fluid and drugs for biomedical system and LOC (laboratory-on-a-chip).

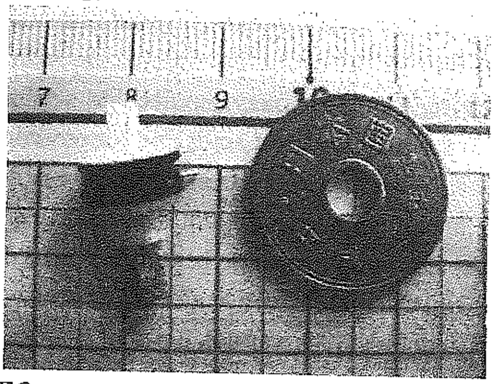

Fig.1 EO actuator used in the diaphragm micropump
(Manufactured by Nano Fusion Technology Inc)

2. Fabrication and Working Principle

As shown in Fig.2, a novel micropump using EO pumps developed in this study comprises three major parts, the EO actuator, the indirect mechanism, the check valve. As shown in Fig.3, a prototype of the micropump was fabricated as the size of Φ30 mm×50 mm (without EO actuators) using acrylic, which is a kind of low cost medical material. The PDMS thin film (the thickness of 50μm, the diameter of 13mm) was clamped between the working liquid chamber and the driving liquid chamber. The width of the clamped PDMS diaphragm edge was 3mm. The two EO actuators which were reversed to each other were installed at the bottom of the micropump. Tow check valves which are able to prevent the working liquid backing flow were installed in the valve seat at the inlet and the outlet. Finally the micropump was sealed to keep it from liquid leakage and contamination of the working liquid. To ensure the high performance of the EO actuators, the DI (deionized) water was used as driving liquid.

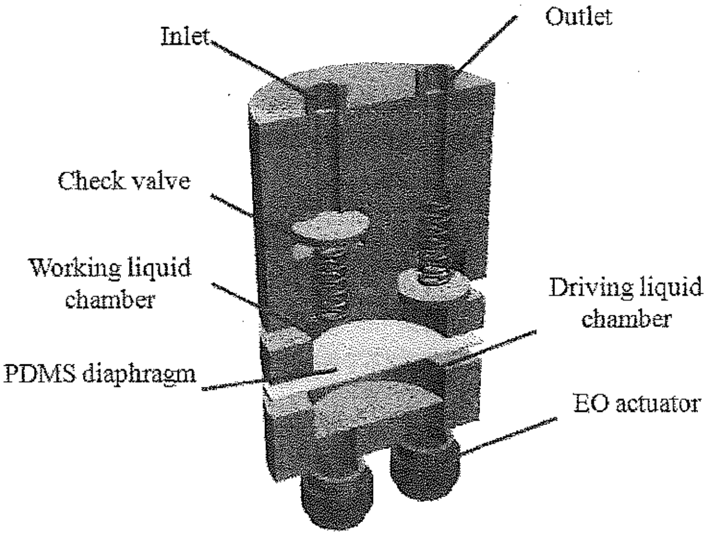

Fig.2 Configuration of the micropump

Fig.4 illustrates the simplified sectional view of the proposed micropump structure and its working principle. A pulsing signal was applied on EO actuators to make them work alternatively. In Fig.4 (a), when the micropump works in suction phase for the suction action of EO actuator, the volume of the driving liquid chamber decreases and the PDMS diaphragm deflects to the downside. And at the same time, the working liquid chamber is getting large in volume and decreasing in pressure. When the pressure get to the threshold pressure, the valve at the inlet will open and the working liquid will flow into the working liquid chamber through the inlet. In the other phase, as shown in Fig.4 (b), the PDMS diaphragm deflects to the upside for the pumping action of the EO actuator. The working liquid chamber is getting small in volume and increasing in pressure. When the pressure get to the threshold pressure, the valve at the out let will open and the working liquid will be pumped out through the outlet. Repeating in this way, the working liquid will continuously flow into the pump's working liquid chamber and flow out through the outlet. If the working liquid flows in the backward direction, the path is closed by the check valve adhering to the valve seat. So the flow can be made in only one direction by the check valves. Besides, the cross-contamination of fluids while it is not activated can be avoided by the micropump with check valves. Moreover, the back pressure is also efficiently enhanced, which is an important parameter for micropumps, because of the normally closed check valves.

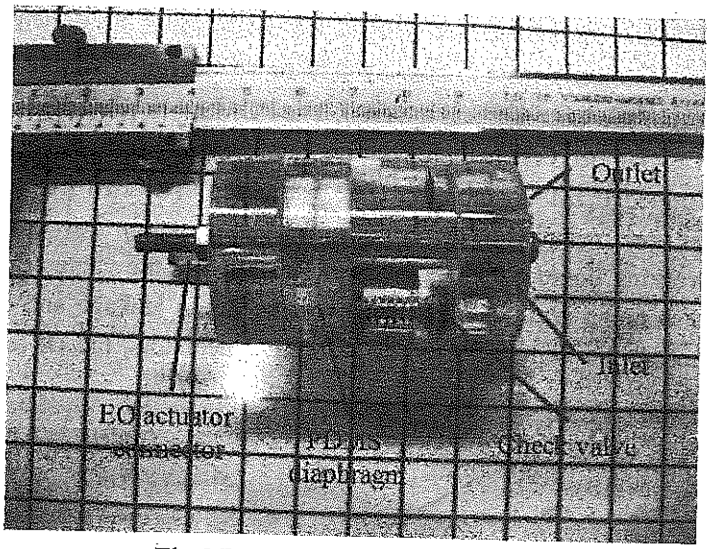

Fig.3 Prototype of the micropump

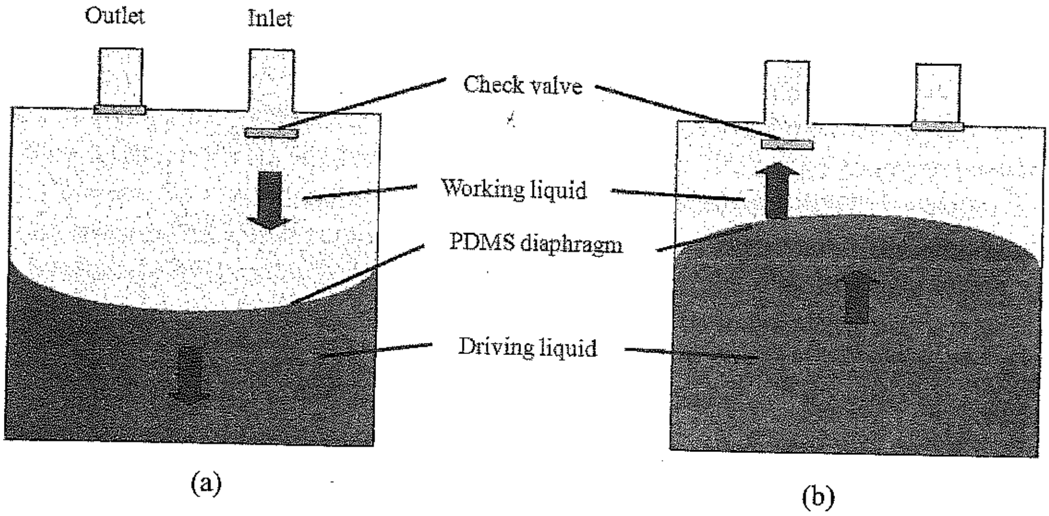

Fig.4 Schematic of the micropump working principle

3. Characterization

3.1 Stroke Volume

In our research, the diaphragm is clamped over an edge of ring area. Since the diaphragm is edge-clamped, the maximum deflection takes place at the center of the diaphragm. CFD (commercial fluid dynamic) analysis programs were used to simulate the deformation of the PDMS diaphragm as a function of electrical power applied to the two EO pumps. In the simulations, fluid in the driving liquid chamber was assumed to be DI water. The output of EO pumps which is also the input of the indirect mechanism was assumed 170μl/min. The ambient are was assumed to be 20°C. As shown in Fig. 5, the biggest pressure is obtained in the center of the diaphragm. And the pressure decreases as the distance to the center increasing. So a parabolic surface can be used to model the deformation of the diaphragm. The parabolic curve can be described as Equation (1).

$$w(r) = -w_{\max}[\frac{1}{R^2}r^2 - 1] \quad (1)$$

Where R is the radius of diaphragm, w_{max} is the maximum deflection of the diaphragm, w(r) is the deflection of the diaphragm at the radius distance of r. Integrating Equation (1) for w(r) over a cross-section gives the stroke volume of the diaphragm in pumping phase.

$$\Delta V_{pumping} = \int_0^{2\pi}\int_0^R w\,(r) r d\theta dr = \frac{1}{2}\pi w_{p-\max}R^2 \quad (2)$$

The stroke volume in suction phase can be calculated by the same method.

$$\Delta V_{suction} = \frac{1}{2}\pi w_{s-\max}R^2 \quad (3)$$

The major design parameter of the mechanical diaphragm type micropumps is called the compression ratio (ε) which is expressed as follows:

$$\varepsilon = \frac{\Delta V}{V_0} = \frac{\Delta V_{pumping} + \Delta V_{suction}}{V_0} \quad (4)$$

Where V_0 is dead volume of the micropump. Substituting Equation (2) and (3) into (4), the relationship between the compression and the maximum deflection of the diaphragm which takes place at the centre of the diaphragm is obtained.

$$\varepsilon = \frac{\pi R^2}{2V_0}(w_{p-\max} + w_{s-\max}) \quad (5)$$

The maximum pressure of the EO actuators generating w_{p-max} and w_{s-max} is given by Zeng [14] as follows:

$$\Delta P_m = \frac{8\varepsilon\varsigma E}{a^2}\left(1 - \frac{2\lambda I_1(a/\lambda)}{a I_0(a/\lambda)}\right) \quad (6)$$

Where $\triangle p_m$ is maximum pressure without any flow. E is the input voltage. ε and ς are dielectric constant and zeta potential of the particles. I_0 and I_1 are the zero-order and the first-order modified Bessel function of the first kind. The average pore radius of the medium is α. But the parameters excluding E are the attribute of the EO actuator and they are constant. So the maximum pressure is in direct proportion to the input voltage.

In order to calculate the compression ratio, an experiment was designed to measure the maximum deflection of the centre of the diaphragm. As shown in Fig.6, the deflection of the diaphragm's centre which is actuated by a pressure chamber connecting to the EO actuators was measured with a laser displacement sensor (KEYENCE LK-2100). To ensure that the diaphragm does not fail during operation, the maximum force applied by pressure chamber must be less than the elastic limit force of the diaphragm. The maximum deflection is 400μm with the applied voltage of 30V. So the compression ratio (ε) calculated is 40%.

3.2 Experiments of check valves

An experiment was designed to study the characteristic of the check valve. As shown in Fig. 6, a pressure difference between the inlet and the outlet made the working liquid flow through the valve. And the flow rate was got by the flow sensor. As shown in Fig 7, when the pressure difference got the threshold pressure (10kPa), the working liquid flowed out through outlet. With the pressure difference increasing slowly, the flow rate became fast.

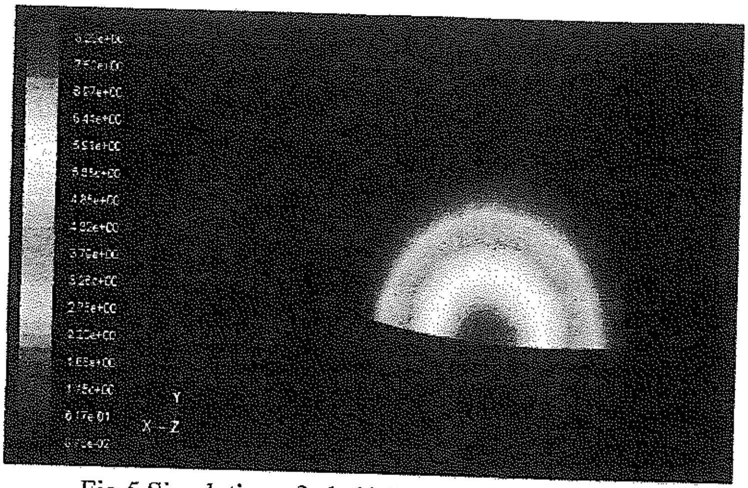
Fig.5 Simulation of a half diaphragm deflection

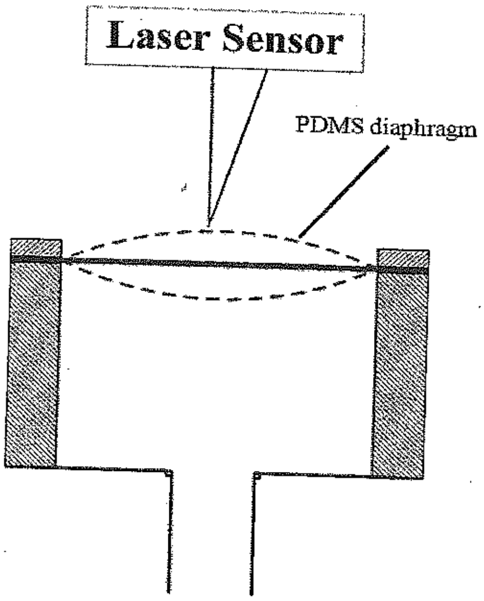

Fig.6 Cross-sectional scheme of the diaphragm actuator

3.3 Measurement of flow rates

The flow rates of the micropump were measured with a flow sensor (LG16-0480A) connecting to the outlet. As shown in Fig 8, through A/D board (PCI-3165) the data of the flow rate was sent to PC. A square wave with a desired frequency and duty ratio which is the input of the driving circuit was generated by D/A (PCI-3329) board. Finally a DC pulse voltage was alternatively applied to the two EO actuators on a desired frequency and duty ratio. The EO actuators were filled with DI water and the indirect mechanism was filled with tap water by syringe. The micro pump was tested using various square pulse signals applied to the EO pumps. Flow rates were measured after a sufficient period of operation and without pressure difference between the inlet and outlet. To ensure that the closed EO pump can cut the flow, the voltage of -1 V which is called gate voltage was applied on the closed EO pump when the two EO pumps working alternatively.

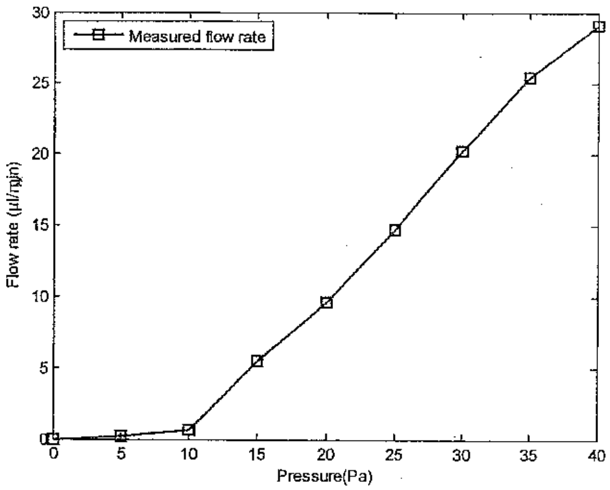

Fig.7 Characteristic of the check valve

3.4 Driving signal

For effective pumping, it is important to find the optimum duty ratio, which is the ratio of pumping time to suction time of the EO actuators. At the optimum duty ratio, the diaphragm is able to deflect from upside-peak to downside-peak before being deformed again with the next pulse. The effect of the duty ratio on the flow rate with the frequency of 1Hz is shown in Fig.9. In general, the flow rate increased with increasing applied voltage. This is because the pressure of the driving liquid chamber increased with an increase in applied voltage, resulting in the diaphragm deforming greatly and an increase in the flow rate. At a duty of 30%, the flow rate was greater than that at a duty of 50% and 70%. And at the optimum duty of 30%, the increase in flow rate was linear with respect to applied voltage. Although the deflection of the diaphragm increased with the time of the voltage-on period for the pulsed voltage getting longer, the diaphragm cannot fully return to the original state or downside-peak at the duty ratio of 50% and 70% for the nonlinear of the diaphragm elastic. In addition, a square-wave is preferred to other signals for the large deflection of the actuator diaphragm. The square-wave input signal at a duty of 30% is used for effective pumping. As shown in Fig. 10, the flow rate was affected by the frequency of the input signal. And the flow rates were measured according to the exciting frequency at a constant duty ratio of 30% for different applied voltage. The flow rate increased with the electrical frequency. However, the flow rate decreased as frequency increased, when it exceed a threshold frequency. The reason of this result is thought to be that at high frequency, the deformation of the diaphragm was not fast enough to get the peak displacement state. A peak flow rate of 133μl/min was received at the frequency of 1.4Hz with the applied voltage of 30V. And when applied voltage is 5V, 10V, 15V, 20V, 25V, the peak flow rates were respectively got at the frequency of 1Hz, 1.1Hz, 1.3Hz, 1.3Hz and 1.4Hz.

3.5 Effects of various factors on flow rates

The flow rate of the proposed diaphragm micropump was influenced by various factors. The most important factor is the characteristics of the diaphragm. In Fig. 11, the flow rates with different thicknesses and different material diaphragm were compared. It is clearly that the micropump fabricated with PDMS diaphragm is optimum to it fabricated with latex diaphragm. So PDMS was chosen as the actuator diaphragm to obtain the biggest output. The flow rates of PDMS diaphragm micropump were investigated by varying the thickness of the diaphragm. With the thickness of the diaphragm increasing, the flow rates decreased. And when the thickness is 200μm, the flow rate almost is under 35μl/min. But if the diaphragm is too thin, it would be broken easily. So the diaphragm must be thick enough so that it is able to support the pressure load. Considering the characteristics of the material and the output flow rate, the optimum thickness for the micropump is 50μm.

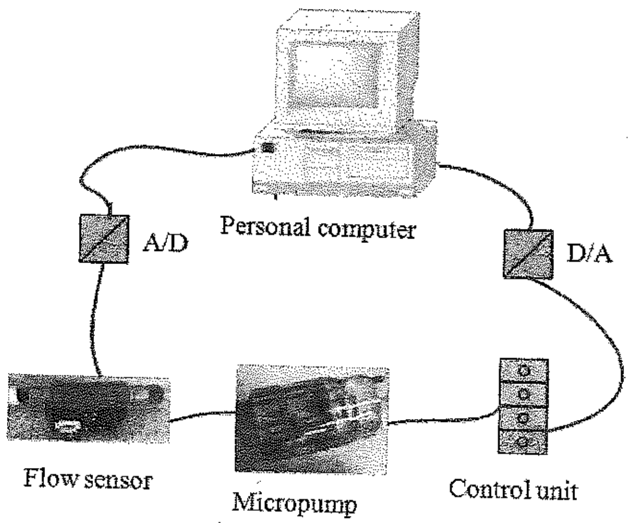

Fig.8 Schematic illustration of the flow rate measurement system.

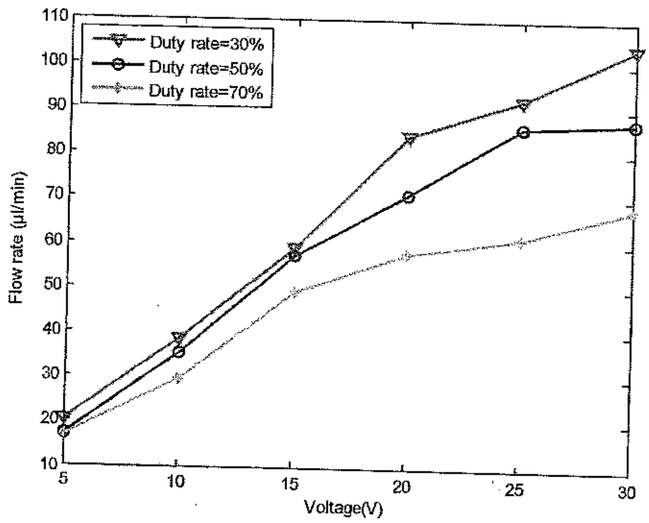

Fig.9 Flow rates with different duty ratios input signal.

(Frequency: 1Hz)

Another factor that could influence the flow rate of the micropump is the check valve. Fig.v12 showed that the flow rate without check valve was obviously smaller than it with check valve. Since the check valves can block the back forward flow and liquid leakage, it takes an important role to enhance the flow rate and output pressure.

In practice, the diaphragm micropump is used with different posture. Fig. 12 compared the micropump flow rates measured in horizontal situation with it with vertical situation. It is only slightly different between these two situations. The peak flow rate of 119μl/min was obtained at the frequency of 1.3Hz with vertical posture. Although it is smaller than that in the horizontal situation for the influence of the gravity, it is able to be accepted in drug delivery system or other biomedical systems. So the micropump could be used in any situations with different postures. Because of this characteristic, it is more suitable for portable bio-medical devices.

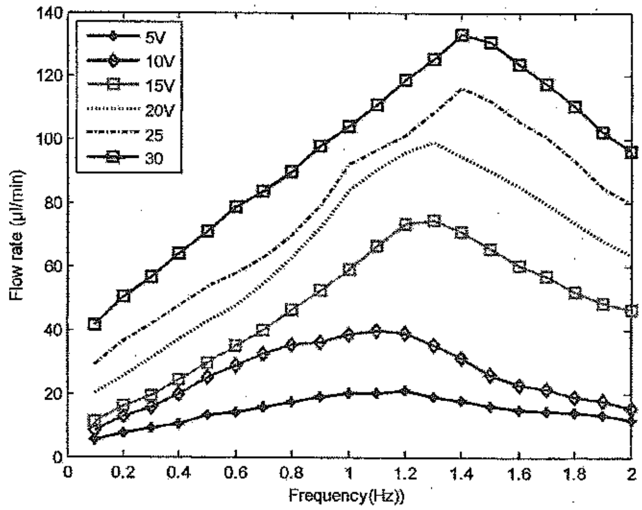

Fig.10 Flow rates with different frequency input signal.

(Duty ratio: 30%, Voltage: 30V)

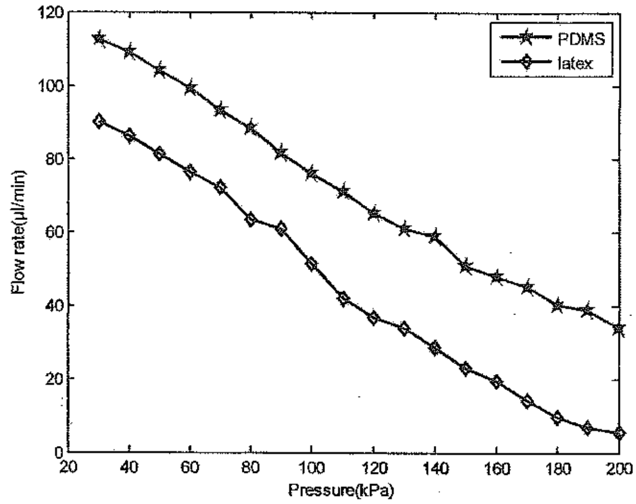

Fig.11 Flow rates with different thicknesses diaphragms

(Duty ratio: 30%, Voltage: 30V)

4. Conclusions

In this paper, a novel PDMS diaphragm micropump based on two EO actuators with micro check valves was designed and fabricated through the rapid prototyping process. The distinct features of the diaphragm micropump include its simple structure by consisting of two sealed chambers, a PDMS diaphragm, two EO actuators and check valves. The flow rate was regulated by controlling the electric frequency and the applied voltage. An optimum duty ratio was found to pump effectively. In addition, the micropump is characterized by low input cost, low input voltage and portability. The dynamic deflection test of the PDMS diaphragm and flow rate test of the micropump were performed. Moreover, considering the simplicity of PDMS diaphragm fabrication process and feasibility of the EO actuation method, it could be applicable to various kinds of drug delivery system and biomedical system.

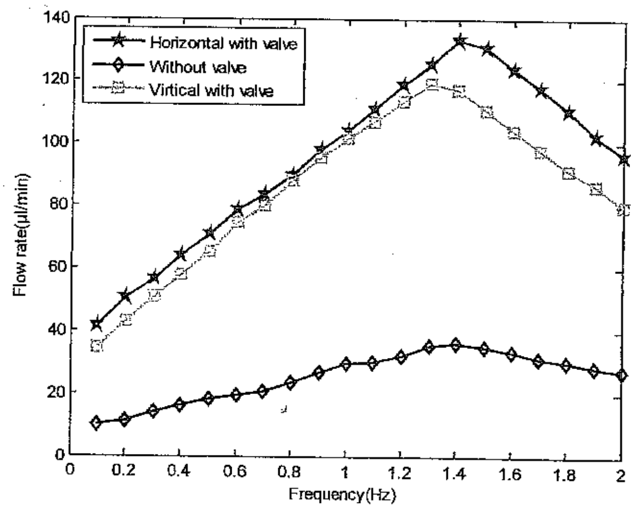

Fig.12 Compare the flow rates in the situation of horizontal with valve, vertical with valve and without valve. (Duty ratio: 30%, Voltage: 30V)

6. Acknowledgement

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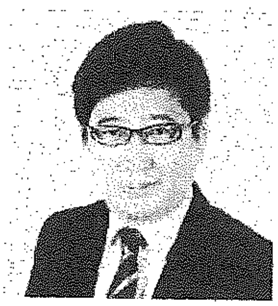

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Dr. Guo received research awards from the Tokai Section of the Japan Society of Mechanical Engineers (JSME), the Tokai Science and Technology Foundation, and the Best Paper Award of the IS International Conference, Best Paper award of the 2003 International Conference on Control Science and Technology, Best Conference Paper Award of IEEE ROBIO2004 and Best Conference Paper Award of IEEE ICAL 2008, in 1997, in 1998, in 2000, in 2003, in 2004 and in 2008, respectively. He is the founding chair for IEEE International Conference on Mechatronics and Automation, and editor in chief for IJMA.

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