A Study on the Passive Microvalve Applicable to Drainage Device for Glaucoma

Tae Seok Sim and Yong-Kweon Kim

Abstract —**This paper reports the design, modeling, fabrication and measurement of passive microvalves, which are applicable to glaucoma implants. The proposed microvalves were designed using fluidic theory. The microvalves consisted of microchannels and chambers. The microchannels had a constant fluidic resistance generating a pressure difference. Six kinds of microvalves were designed using fluidic equations for laminar flow and fabricated to examine the influences of chamber size, channel length and the shape of channel cross section. The pressure difference between the designed microvalve and the fabricated microvalve was measured to be less than 4%.**

Index Terms— **MEMS, Micro fluidics, Glaucoma, Passive, Microvavle**

I. INTRODUCTION

Glaucoma is an eye disease caused by abnormally high eye pressure resulted from a poor drainage of the eye fluid. It damages optic nerves and leads to visual loss and blindness. Currently, an implantation surgery is known to be the most effective treatment for the incurable glaucoma patients. However, conventional drainage implants have serious problems such as a low fluidic resistance and a big size. The low fluidic resistance causes excessive drainage of the eye fluid after a surgical treatment, and this leads to a hypotonicity (low intra ocular pressure state). The big size may give a fear to patients before surgical treatments [1]. The problems described above would be removed if the glaucoma implant could be fabricated to be more accurate and small [2,3].

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School of Electrical Engineering and Computer Science, Seoul National University

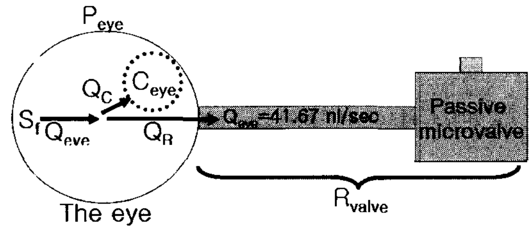
In this paper, passive microvalves using fluidics and MEMS technology were proposed. The proposed microvalve consists of microchannels and a chamber.

Modeling, design, fabrication and measurement of the proposed microvalve have been described and the optimum microvalve for glaucoma is discussed by comparing the measured results.

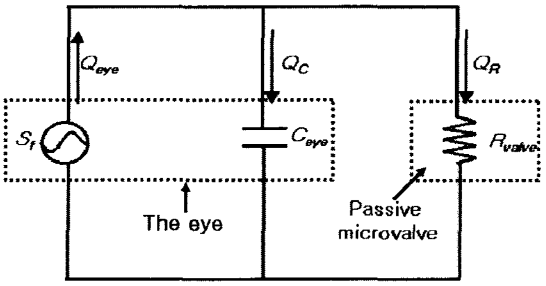
II. THEORY

A. *Modeling of valve implanted eye system*

An eye fluid, called aqueous humor, is produced in a posterior chamber of an eye from circulating blood with a fixed rate, 4.167 nl/sec. Thus the eye can be assumed as a universal flow source. The cornea-sclera envelop expands and relaxes according to the variations in the internal volume of an eye ball. Therefore the eye can be assumed to have a compliance factor. Let P_{eye} be the pressure of eye ball, C_{eye} be the compliance of eye ball, S_f be the flow source, R_{valve} be the fluidic resistance of the valve, Q_{eye} be the flow rate of eye ball and Q_{valve} be the flow rate of channel. The valve implanted eye system can be simplified as Fig. 1. (a). Since pressures, flow rates, fluidic resistances and compliances of the fluidic systems have analogies to voltages, currents, resistances, capacitances of the electrical systems each other. The valve implanted eye system can be depicted by an



(a) Schematic of the valve implanted eye system



Q_{eye}	R_{valve}	P_{eye}	C_{eye}
i	R	V	C

(b) An equivalent electrical circuit

Fig. 1. Schematic of the valve implanted eye system and an equivalent circuit.

equivalent electrical circuit as shown in Fig. 1. (b).

From the equivalent electrical circuit (Fig. 1. (b)), we can formulate the following equations,

$$Q_{eye} = Q_C + Q_R = C_{eye}\frac{dP_{eye}}{dt} + \frac{P_{eye}}{R_{valve}} \tag{1}$$

Solving this differential equation, we obtain

$$P_{eye}(t) = (P_{eye}(0) - R_{valve}Q_{eye})e^{-\frac{t}{R_{valve}C_{eye}}} + R_{valve}Q_{eye} \tag{2}$$

In the Eq. (2), if t goes to infinity, the first term becomes zero. Thus, for the steady state, the eyeball pressure is expressed by the following simple equation [4, 5],

$$P_{eye}(t) = R_{valve}Q_{eye} \tag{3}$$

In the Eq. (3), Q_{eye} is an uniformly fixed value. Hence the eyeball pressure is controlled by only fluidic resistance of the valve. Since the normal intra ocular pressure (IOP) is ranged from 1,000 Pa to 3,000 Pa and the average is 2250 Pa, 2000 Pa seems to be the most proper and generous target pressure for a prototype microvalve. However, the fluidic resistance becomes gradually large due to a fibrosis in practice. Thus, if the fibrosis is considered during the design procedure, the target pressure should be smaller than 2000 Pa. Therefore 1000 Pa (minimum normal intra ocular pressure) was set for the prototype pressure difference.

B. *Design of the passive microvalve*

Before the design of the passive microvalve, the flow type was defined using Reynolds number because the equations used for fluidic resistance are different according to the flow type. The equation of Reynolds number is expressed as,

$$\mathrm{Re} = \frac{\rho VD}{\mu} \tag{4}$$

Where ρ: fluid density of water (kg/m^3), V: average velocity of the fluid in the pipe (m/s), D: pipe radius (m) and μ: dynamic viscosity (N·s/m^2). Since the flow rate is fixed and already known to be 41.67 nl/sec, the average velocity is determined by the channel cross section size. Among the various available choice of the channel cross section, rectangular channel cross section of 80 x 80 μm^2 was chosen for the standard channel cross section and the acquired average velocity was 0.00651 m/s. With the chosen cross section size, the calculated Reynolds number was 0.521 and the flow type was a laminar flow. In the laminar flow system, the fluidic resistance for arbitrary shape is expressed as,

$$R = \frac{f\,\mathrm{Re}}{2}\frac{\mu l}{D_h^2 A} \tag{5}$$

In the Eq. (5), fRe, μ, l, D_h and A are shape constant for arbitrary non-circular channels, absolute viscosity of water, channel length, hydraulic diameter and area of the channel cross section, repectively. D_h can be calculated by the following equation.

$$D_h = 4\frac{Area}{Perimeter} = \frac{2ab}{(a+b)} \tag{8}$$

where a and b are the length of sides of the rectangular channel cross section. fRe is given as,

$$f\,\mathrm{Re} = 96(1 - 1.3553\alpha + 1.9467\alpha^2 - 1.7012\alpha^3 + 0.9564\alpha^4 - 0.2537\alpha^5) \qquad (9)$$

where,

$$\alpha = \frac{b}{a}, \text{ in case of } a \geq b \qquad (10)$$

Since the target pressure difference was already known, the channel length was calculated using Eq. (5). The design layouts are shown in Fig. 2 and the design parameters are summarized in Table 1.

Models ①, ② and ③ were designed to have different channel cross section size for the same target pressure difference to examine the feasibility of the Eq. (5) for various channel cross section sizes. Models ①, ④ and ⑤ were designed to have different chamber size to examine the influence of chamber. Finally, model ⑥ was designed to have the target pressure difference of 2,000 Pa to examine the relation between the fluidic resistance and the channel length.

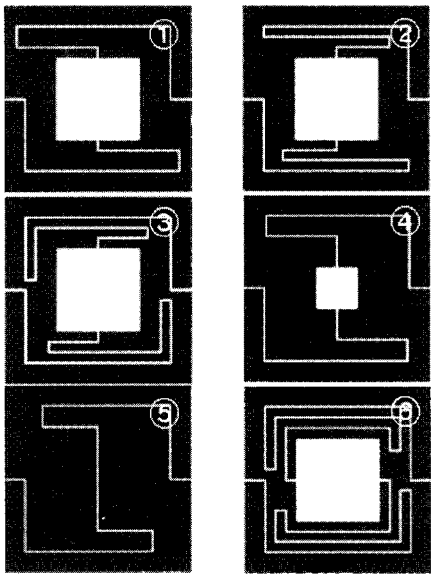


Fig. 2. Six kinds of design layouts.

Table 1. Design parameters.

Model #	Channel cross section (μm²)	Channel length (mm)	Chamber Size (mm²)	Target pressure difference (Pa)
①	80×80	34.6	4×4×0.08	1000
②	90×80	43.3	4×4×0.08	1000
③	100×80	52.5	4×4×0.08	1000
④	80×80	34.6	2×2×0.08	1000
⑤	80×80	34.6	none	1000
⑥	80×80	69.2	4×4×0.08	2000

III. FABRICATION

The fabrication procedure of the proposed passive microvalves is depicted in Fig. 3.

Thermal SiO_2 was patterned to define a hard mask. Then Si was etched to a depth of 80 ㎛ by DRIE (Deep Reactive Ion Etching) with the patterned hard mask. DRIE was performed using 535 bosch process in an ICP reactor (Plasma Therm SLR-10R-B). The inlet and outlet of the microvalve were defined by dicing process.

Finally, anodic bonding was done to make the channel with a glass wafer. The fabricated valve is shown in Fig. 4. The dimensions of the fabricated microvalves and the

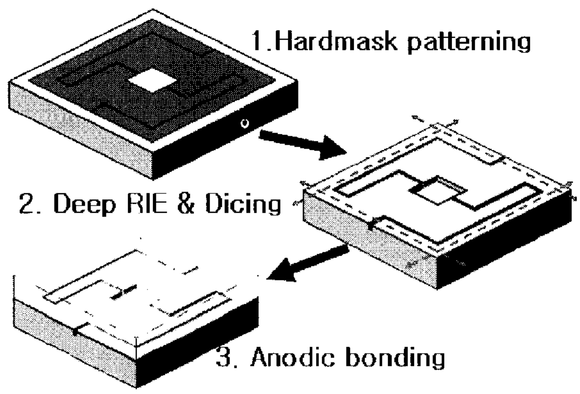


Fig. 3. Fabrication process of the proposed microvalve.

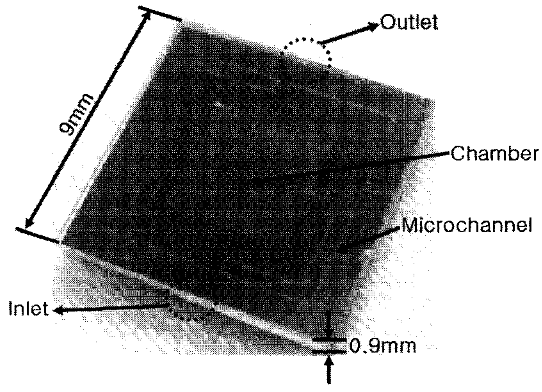


Fig. 4. The fabricated microvalve.

expected pressure differences are summarized in Table 2. Due to the fabrication error, there are disagreements between the expected pressure differences and the designed target pressure differences. However, the amounts of the target pressure changes are relatively small compared to the designed target pressure differences. The maximum percentage of the error was 4.2%.

IV. MEASUREMENTS & RESULTS

A. *Measurement system*

The schematic of measurement system is depicted in Fig. 5. The microvalve was filled by deionized water to prevent air bubble generating in the middle of microvalve before measurement. A syringe pump was used to supply deionized water to the microvalve at a rate of 41.67 nl/sec. A pressure sensor was connected between the syringe pump and the microvalve. Data from the pressure sensor were acquisited using NI-DAQ, PCI-MIO-16E-1 board and LabVIEW program. The pressure measurement was achieved five times for each model.

B. *Measurement results*

The measured pressures are summarized in Table 2. All the average pressure differences showed good agreements with the expected target pressures. The maximum difference between the average pressure and the expected pressure was about 25 Pa (2.5% of the expected pressure difference). The results of model ① (target pressure: 1000 Pa) and model ⑥ (2,000 Pa) showed that the pressure difference through the microvalve is proportional to the that of channel length. For various channel cross section sizes (comparison of models ①, ② and ③), all the measured results were near the expected pressure difference. In the case of comparison of chamber size (models ①, ④ and ⑤), the measured results were also near the expected pressure difference. However, it was observed that the bigger the chamber size becomes, the slower the response time becomes (Fig. 6). The chamber was designed for movable membrane, if the valve is driven actively.

The time responses are less than 50 sec and they are negligible for glaucoma implants.

The pressure difference of model ① was measured for about 9.17 hours to examine the stability of the microvalve. The measured result is shown in Fig. 7. It

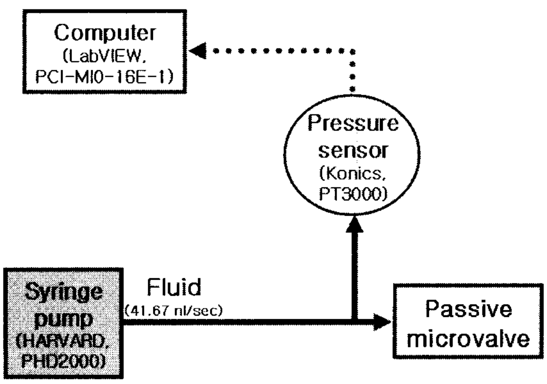


Fig. 5. A schematic of measurement system.

Table 2. Summary of the fabrication results and the measured pressures.

#	Designed Pressure Difference (Pa)	Fabrication results			Pressure measurement results
		channel cross section (μm^2)	channel length (mm)	Expected pressure difference (Pa)	Average ± σ (Pa)
①	1000	80.6×79.8	34.6	992	1007 ± 5
②	1000	90.5×81.0	43.3	961	963 ± 6
③	1000	99.8×81.0	52.5	972	978 ± 7
④	1000	80.7×81.0	34.6	960	982 ± 6
⑤	1000	81.4×80.4	34.6	958	970 ± 8
⑥	2000	80.7×80.4	69.2	1950	1963 ± 11

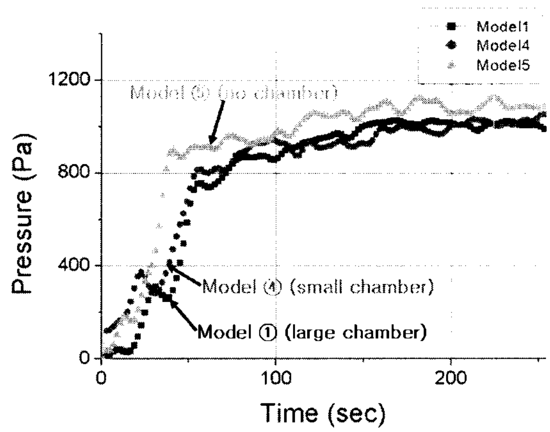


Fig. 6. Pressure difference comparison of model ①, ④ and ⑤.

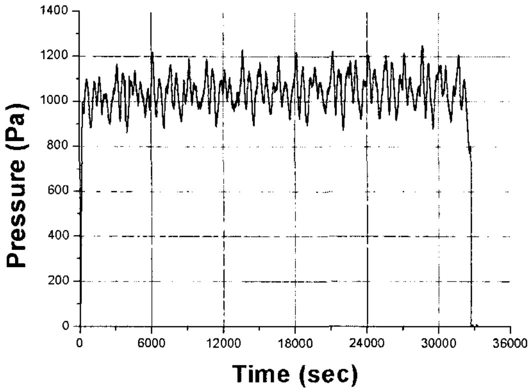


Fig. 7. Pressure difference of the model ① for 9.17 hours.

was observed that the fabricated microvalve successfully operated without pressure drift for about 9.17 hours. In the Fig. 7, data fluctuations resulted from a nonuniform flow rate due to a stepwise motion of the syringe pump in the measurement system.

V. DISCUSSION

During the design procedure and the calculation of the pressure differences, minor losses such as 90° bend of pipe, sudden expansion and sudden contraction were neglected (Fig. 4). In the laminar flow, the pipe bending is usually neglected, because of the very slow average velocity of the fluid. In the case of sudden expansion and sudden contraction, the loss coefficient, K, can be expressed as Eq. (9) and Eq. (10).

$$K_{\text{sudden expansion}} \approx (1 - \frac{d^2}{D^2})^2 \quad (9)$$

$$K_{\text{sudden contraction}} \approx 0.42(1 - \frac{d^2}{D^2}) \quad (10)$$

D is the diameter of large channel and d is the diameter of small channel. the calculated loss coefficients of sudden expansion and sudden contraction were 1 and 0.42, respectively. The pressure differences generated by sudden expansion and sudden contraction can be calculated using the Eq. (11).

$$\Delta P = K \cdot \frac{1}{2} \rho V^2 \quad (11)$$

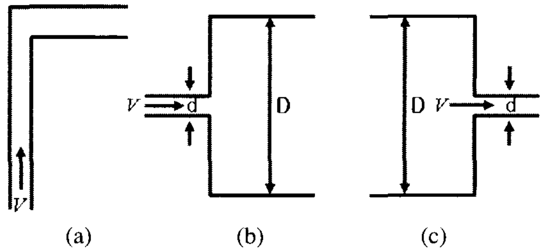


Fig. 7. Several types of minor losses that may be seen in the proposed microvalves. (a) 90° bending, (b) sudden expansion, (c) sudden contraction.

where ρ is the fluid density and V is the average fluid velocity. The calculated pressure differences were 0.0212 and 0.0089 Pa (<< 1 Pa). Thus the pressure differences due to the minor losses may be negligible.

The influence of the channel cross section size can be explained as follows. Reforming the Eq (5) for a and b, the fluidic resistance can be expressed as,

$$R(a,b) = \frac{f\,\mathrm{Re}}{2} \frac{\mu l}{D_h^2 A} = \frac{f\,\mathrm{Re}\,\mu l}{8} \frac{(a+b)^3}{a^3 b^3} \quad (12)$$

Assume that there is a small change in the one side of the channel cross section. Then the changed fluidic resistance can be expressed using sensitivity method and linearization as shown below,

$$\frac{\Delta R}{R} = (\frac{2}{a+b} - \frac{3}{a}) \Delta a \quad (13)$$

In the case of a≈b,

$$\frac{\Delta R}{R} = -2 \frac{\Delta a}{a} \quad (14)$$

From the Eq. (14), if the channel cross section is large, the rate of pressure change becomes small, because the fabrication error due to the micromachining technology does not depend on the feature size. The fabrication error was less than 1μm and the influence on the pressure difference is less than 2%. Thus It was found that it is better to design the channel cross section size as big as possible.

VI. CONCLUSION

The passive microvalves applicable to glaucoma implant have been fabricated and investigated, using the described design method and MEMS technology. All the measured pressure differences of the fabricated microvalves showed good agreement with the target pressure differences and it was also shown that the fabricated microvalve operated well without pressure drift for approximately 9.17 hours. Consequently, fabrication of the passive microvalve for glaucoma seems to be possible. For future works, in vivo test is needed to examine the feasibility of the microvalve in practice.

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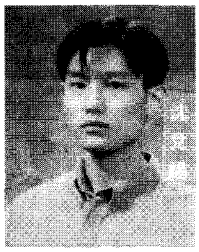
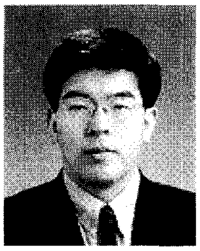


Tae seok Sim received the B. S. and M. S. degrees in electrical engineering from Korea University, Seoul, Korea in 1999 and 2001, respectively, and He is currently a Ph.D. candidate at School of Electrical Engineering and Computer Science in Seoul National University, Seoul, Korea. His main recent research activities are micro fluidics and microvalve.



Yong-Kweon Kim (S' 90-M' 90) received the B.S. and M.S. degrees in electrical engineering from the Seoul National University, Seoul, Korea, in 1983 and 1985, respectively, and the Dr. Eng. Degree from the University of Tokyo, Tokyo, Japan, in 1990. His doctoral dissertation concerned modeling, design, fabrication, and testing of microlinear actuators in magnetic levitation using high critical temperature superconductors.

In 1990, he joined the Central Research Laboratory, Hitachi Ltd., Tokyo, Japan, where he was a Researcher involved with actuators of hard disk drives. In 1992, he joined the Seoul National University, where he is currently an Associate Professor of electrical engineering. His current research interests are modeling, design, and fabrication and testing of electric machines, especially MEMS, microsensors, and actuators