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Design and simulation of a novel electrostatic peristaltic micromachined pump for drug delivery applications

Mir Majid Teymoori*, Ebrahim Abbaspour-Sani

Urmia MEMS Lab, Department of Electrical Engineering, Urmia University, Beheshti Ave., Urmia 57135, Iran

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Abstract

A novel electrostatic micromachined pump for medical applications is designed and simulated. The proposed structure for the micropump consists of an input and an output port, three membranes, three active membrane valves, microchannels, and three electrostatic actuation systems. Pumping mechanism of the proposed micropump is based on the peristaltic motion that has some advantages, such as high controllability and precision, over the other mechanism that makes it suitable to be used for the medical applications. Electrostatic actuation has been employed for the deflection of the membranes because of its benefits, such as the smaller size of the device in comparison with the other types, especially piezoelectric counterpart and so on. Employing active membrane valves instead of passive check valves resolves some of the problems, such as valve clogging and leakage. The designed micropump satisfies all medical drug delivery requirements, such as drug compatibility, flow rate controllability, self-priming, small chip size, and low power consumption. The flow rate of the designed micropump is 9.1 μl/min which is quite suitable for drug delivery applications, such as chemotherapy. Total size of the designed micropump is 7 mm × 4 mm × 1 mm, which is smaller than the other peristaltic counterpart micropumps. Assuming zero residual stress, low actuation voltage, and small size are the main advantages of our design. The designed micropump is simulated by the finite element method, using the ANSYS 5.7 software.
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Keywords: MEMS; Micropump; Drug delivery; Electrostatic; Peristaltic; Finite element

1. Introduction

Micromachining technology is the miniaturization technique for manufacturing devices and systems, using integrated circuit (IC) fabrication process, such as etching, bonding and so on. Micro electro mechanical system (MEMS) has been opened new thrusts into the world and make it possible to fabricate small size devices and systems with high functionality, precision, and performance. Based on these characteristics, MEMS devices and systems have found some applications, such as automobile, aerospace, communication, medical, etc.

A number of medical devices and systems, such as blood pressure sensors, microneedles, glucose sensor, DNA analyzing system, etc., are designed and fabricated[1,2].

Micropump is one of the MEMS devices, which can be used for drug delivery applications. This device as the main part of a drug delivery system transfers the fluid (drug) from the drug reservoir to the body (tissue or blood vessel) with high performance, accuracy, and reliability. Small size and high precision of micropumps have made them useful for chemotherapy, insulin delivery for diabetic patient, and drug dosing for cancer patient and so on [3].

In this paper, the theory of the pumping mechanism and actuation method for drug delivery micropumps are discussed. The novel structure of our design and its simulation results are also presented, which show good compatibility with the drug delivery requirements.

* Corresponding author. Tel.: +98 441 3457455; fax: +98 441 3454842.
E-mail addresses: ms0062@mail.urmia.ac.ir (M.M. Teymoori), e.abbaspour@mail.urmia.ac.ir (E. Abbaspour-Sani).

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2. Drug delivery micropumps

Micropumps, like the advanced micromachined devices, are the best candidates for drug delivery application. A number of medical micropumps based on different actuations and pumping mechanisms have been designed and fabricated [4-6].

Drug delivery micropumps must satisfy seven important requirements, which are [7]: drug compatibility, actuation safety, flow rate, self-priming, chip size, controllability of flow rate at all times, and power consumption.

A drug delivery device (micropump) must not introduce any toxic particles into the drug and vice versa [8]. Furthermore, the actuation mechanism of micropump must not damage and electrolyze the drug. Since, the magnetic field and thermopneumatic actuation can affect the drug quality [9,10], mechanical micropumps, such as piezoelectric and electrostatic types are suitable for this purpose.

The flow rate of the drug delivery micropump must be in the range 10–100 μl/min. However, the general flow rate of the medical micropumps is considered 10 μl/min [11]. It should be noted that the flow rate of medical micropumps must be controlled at all times.

Self-priming characteristic, which describes the ability of a micropump to fill itself completely with liquid without any additional measures, can be achieved by large stroke and small dead volume.

For the medical applications, the chip size and power consumption of the micropump should be reduced.

3. A novel micropump: design

Based on the medical parameters mentioned in the previous section, we propose a novel structure for the micromachined pump. The three-dimentional structure of the micropump is shown in Fig. 1a.

As shown in the Fig. 1b, our structure consists of three layers: glass, Si substrate, and membrane part. Membrane part includes three active valves on top of the membranes, microchannels, three electrostatic chamber (air gap), input, and output ports, which are schematically shown in Fig. 1c. Working principle of the proposed micropump is based on the peristaltic motion which is schematically shown in Fig. 2 and has some advantages over the other pumping mechanism, such as individual actuation of each membrane. The scalability, fast mechanical response, bi-directional flow capability, and low power consumption of the electrostatic actuation enforce us to employ it for our designed micropump.

Active membrane valve is used for flow–stop capability of our micropump, which is depicted in Fig. 3. This type of valve is firstly used by Cao et al. [12] and do not have the problems of the two-passive check valves, such as valve clogging, breaking, and leakage. As shown in the Fig. 3, active membrane valve consists of two parts: membrane and sealing part, which is attached to the membrane. This part is normally fixed on the input and output ports and provides normally closed conditions for micropump. The active part deflects as soon as membrane is deflected.

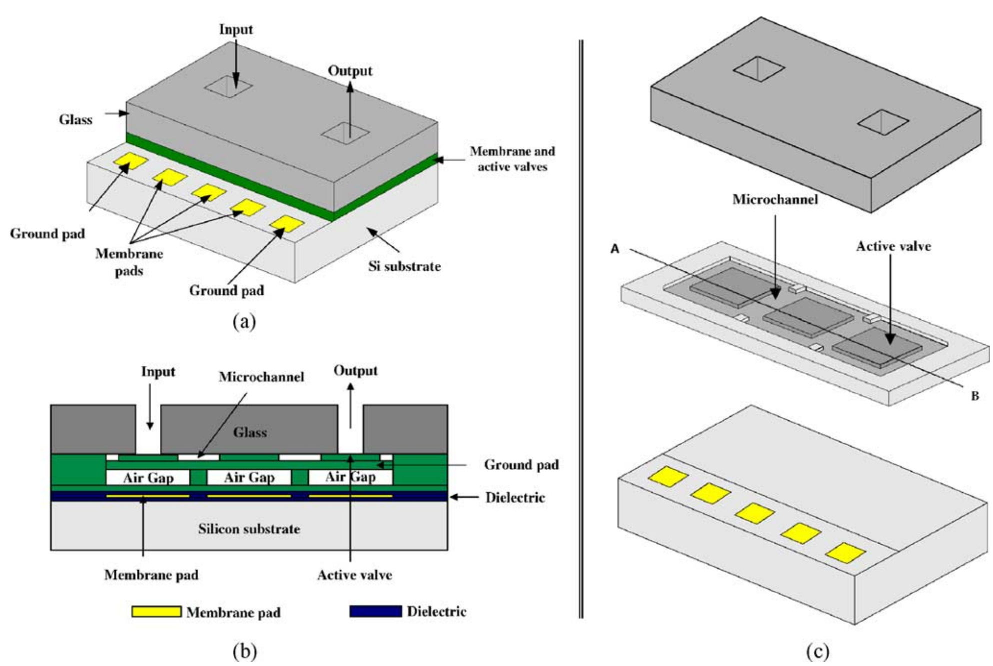

Fig. 1. (a) Three dimentional schematic of the proposed micropump (glass (■); Si substrate (□); gold pads (□); membrane part (■)). (b) Three layers of proposed structure (glass (■); Si substrate (□); gold pads (□); membrane (□); microchannel (□)). (c) Cross-section of the proposed micropump (A–B view) (glass (■); Si substrate (□); gold pads (□); dielctric (■); membrane (■)).

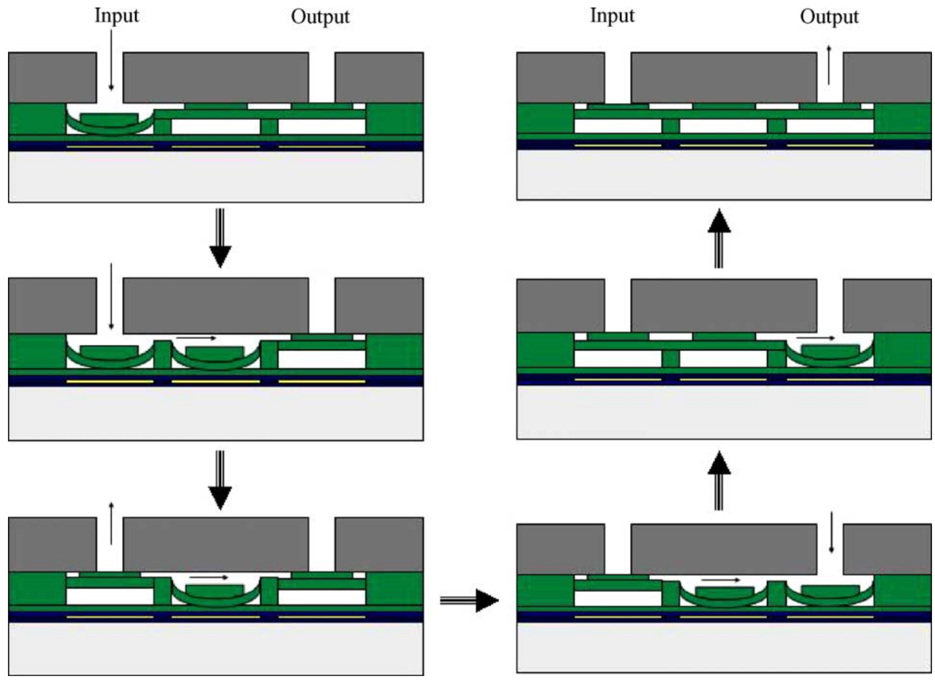

Fig. 2. Working principle of proposed micropump (glass (■); Si substrate (□); gold pads (□); dielctric (■); membrane (■)).

Since Cao et al. [12] had used piezoelectric discs for the actuation, total size of the micropump was very large. However, our designed micropump employs the electrostatic actuation, so we can decrease the size of membranes to desired value and achieve small size micropump.

Our designed micropump satisfies all medical parameters mentioned in the previous section. The suggested materials for fabrication are: polysilicon, glass, SiO_2, Au and Si_3N_4, which have chemical stability and drug compatibility characteristics. It also employs the active membrane valves for flow handling and controlling. Since, the electrostatic field is applied outside the microchannels, the hydrolysis problems are prevented. Dead volume of our proposed structure is very small and the compression ratio (stroke volume/dead volume) is almost 0.8 which guarantees the self-priming capability of the designed micropump. Total size of our designed micropump is 7 mm × 4 mm × 1 mm, which is smaller than the other peristaltic micropumps.

4. Analytical considerations

4.1. Static analysis

Suppose that an external voltage (V) is applied between two electrodes, the first one is fixed and the other is a thin and flexible film. The relationship between the applied voltage and the mechanical spring constant is expressed as [13]:

$$V_{\mathrm{th}} = \sqrt{\frac{8Kg_0^3}{27\varepsilon A}} \quad (1)$$

where A is the effective plate area, V_{th} is the threshold voltage, g_0 is the gap between two electrodes, ε is the permittivity and K is the mechanical spring constant. For our design, A = 1.5 mm × 1.5 mm; g_0 = 4 μm, and ε = 8.854E−12. To reduce the threshold voltage, which is one of our aims, the best way is to reduce the spring constant of the membrane (K).

To determine V_{th}, the mechanical spring constant must be calculated. The governing mechanical equation for a completely all sides-fixed rectangular membrane with uniform thickness is [14,15]:

$$D_0h^3\Delta g(x, y) - T_0h\Delta g(x, y) = p(x, y) \quad (2)$$

where:

$$D_0 = \frac{E}{12(1-\nu^2)} \quad (3)$$

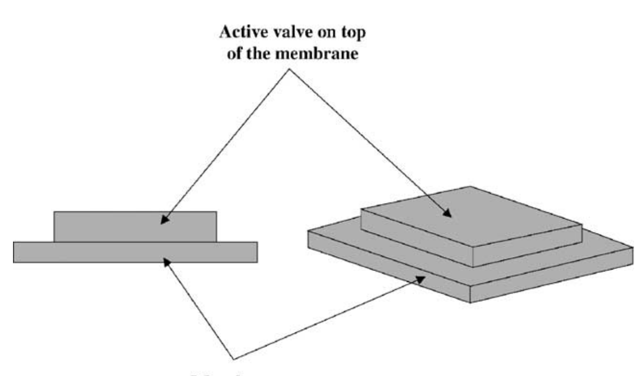

Fig. 3. Active membrane valve (membrane (□)).

$$P(x, y) = \frac{P_0}{(1 - (w(x, y)/g_0))^2} \quad (4)$$

$$P_0 = \frac{\varepsilon V^2}{2g_0^2} \quad (5)$$

and $\Delta = (\partial^2/\partial x^2) + (\partial^2/\partial y^2)$, where D_0 is the mechanical rigidity, E the Young's modulus of the membrane, ν the Poisson's ratio, T_0 the residual stress on the membrane, $w(x, y)$ the deflection of the membrane, ε the air permittivity, g_0 is the gap between two electrodes, and $p(x, y)$ the electrostatic pressure. Solving Eq. (2), the spring constant can be calculated as below:

$$K = 66\frac{Eh^3}{a^2(1 - \nu^2)} \quad (6)$$

As shown in the Eq. (6), the mechanical spring constant of the membrane depends on:

(a) Material property of the membrane (E: Young's modulus, ν: Poisson's ratio)
(b) Membrane dimension (a: length, h: thickness)
(c) Membrane's side degrees of movement
(d) Load position

Due to good mechanical and electrical characteristics of the polysilicon, such as high yield strength, it is selected as the membrane material [16]. Since, we prefer to design a small size micropump, therefore, it is not a good idea to increase the dimensions of the membrane for decreasing the V_{th}. Furthermore, reduction in the thickness of the membrane will decrease its lifetime. Therefore, the only remaining possibility is to increase the membrane degrees of freedom to decrease the threshold voltage. So, we proposed the "Two sides completely fixed and the other two sides semi-free" structure for our design as shown in Figs. 4 and 5.

Classic solution of Eq. (2) with our initial movement conditions is almost impossible. However, there are numerical and finite element methods (FEM) to calculate its mechanical spring constant. Since, the degrees of freedom of our proposed structure is increased, therefore, the spring constant of it has decreased, which is our main goal. Therefore, the threshold voltage of our structure is smaller than the similar structure with all sides fixed.

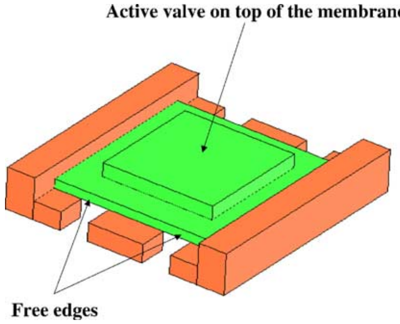

Fig. 4. Membrane edges degrees of freedom (membrane (■); membrane support (■)).

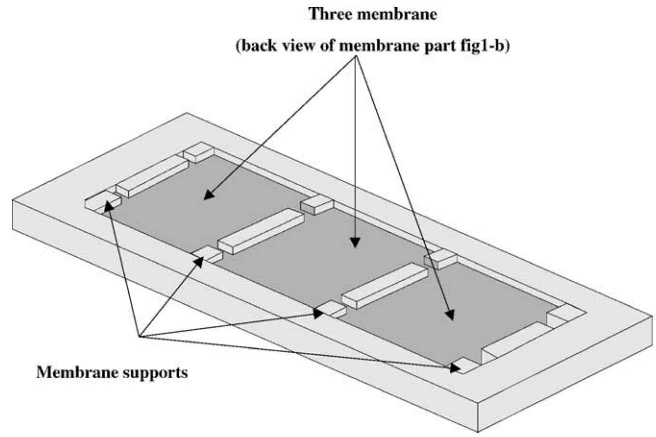

Fig. 5. Degrees of freedom of three membranes (membrane (■); membrane support (■)).

4.2. Dynamic (frequency) analysis

To evaluate the dynamic response of the membranes, transient analysis is necessary. The governing dynamic equation for a completely fixed membrane with voltage (V) applied between the membrane and the fixed electrode is:

$$m\frac{\partial^2 Z}{\partial t^2} + b\frac{\partial Z}{\partial t} + KZ = \frac{\varepsilon_0 A V_s^2}{2g_0^2} \quad (7)$$

with the initial conditions of: $Z(t = 0) = 0$, $(\partial Z(t = 0)/\partial t) = 0$where m is the effective mass of the membrane, b the damping coefficient and V_s is the applied voltage. Assuming $b = 0$ (no damping), $Z(t)$ is calculated as:

$$Z(t) = \frac{\varepsilon_0 A V_s^2}{2Kg_0^2}(1 - \cos(\omega_0 t)) \quad (8)$$

where $\omega_0 = \sqrt{k/m}$

Applying the conditions of $Z(t = t_s) = g_0$ gives the switching speed as:

$$t_s = 3.65\frac{V_{th}}{V_s\omega_0} \quad (9)$$

where V_{th} is the threshold voltage and ω_0 the resonant frequency of the structure.

Sometimes it is necessary that the flow rate of micropump must be greater than our designed flow rate. To achieve a higher flow rate, the best way is increase the pumping frequency. Generally, two factors affect the high frequency applications of electrostatic actuators:

(1) Natural frequency of the structure,
(2) Damping force of the electrostatic chamber.

Each structure has natural frequency, which limits its actuation frequency. Air damping force can also affect the actuation frequency of the electrostatic actuators. To resolve this problem, usually some holes are created on the membrane

[17]. The air is passed through these holes during the membrane deflection and the resistant energy is reduced against the actuated energy. Because of the medical application of our designed micropump, creating holes on the membrane is not possible. However, we can use the bottom electrode and the sacrificial layer removal holes for this purpose. This method reduces the air damping force considerably and our micropump can be used frequencies as high as 100–150 Hz (based on simulation results).

Based on Eqs. (8) and (9), to achieve low switching time (pumping time in our case) and consider low actuation voltage, ω_0 must be increased. However, a trade-off between low actuation voltage and high pumping frequency must be achieved.

To increase both the switching speed (pumping frequency here) and decrease the threshold voltage, V_s is selected between 1.2 and 1.5 times greater than V_{th}.

Eq. (9) is true, only for electrostatic structures, which act in the vacuum ambient ($b = 0$). Since the output port of our micropump is connected to the body (tissue or blood vessel), normally to keep the valve closed condition, the pressure of the electrostatic chambers must be greater than the blood pressure and the vacuum conditions cannot be true for our proposed micropump. So, we must consider the damping effect on the flow rate of our micropump.

It should be noted that the applied voltage is usually larger than the calculated analysis. The main reason is the residual stress on the movable structure (membrane). This stress can even increase the applied voltage three times larger than the calculated one. There are two main factors for creation of residual stress in our structure:

(a) Temperature cycling of the process, especially in the membrane part,
(b) Bonding process.

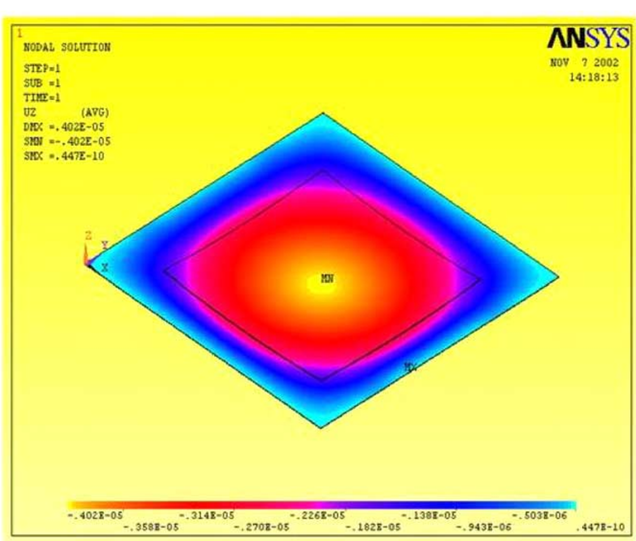

Fig. 6. Displacement of our structure induced by electrostatic force (18.5 V).

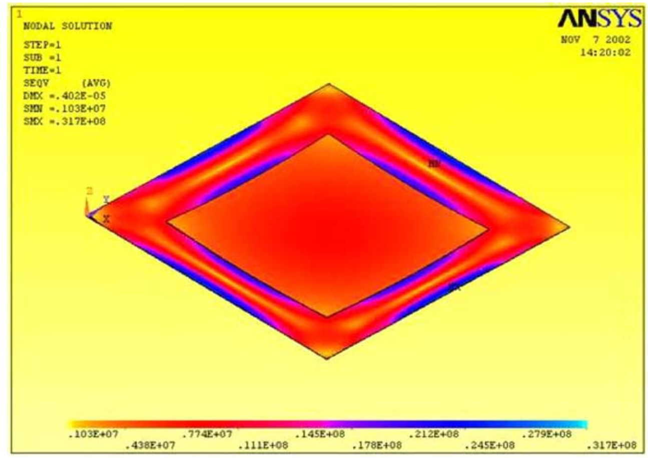

Fig. 7. Stress distribution of our structure induced by electrostatic force (18.5 V).

5. Simulation

To evaluate the threshold voltage and dynamic response of the proposed micropump, the structure is simulated employing finite element method with ANSYS software. The voltage is applied between the membrane and the fixed electrode, and then its behavior is investigated. Two types of membranes are considered in this simulation: One, with all edges clamped and the other with two edges clamped while the other two edges are semi-free (Fig. 4). The main aim of this simulation is to compare the two structures with varying stiffness (different spring constants K). The membrane size for both structures is 1500 μm × 1500 μm × 2 μm, with sealing part, 1000 μm × 1000 μm × 3 μm, and the electrostatic gap is taken 4 μm. To simplify, in all simulations it is assumed that the residual stress on the membrane is zero. It is evident that residual stress will increase the threshold voltage.

The pull-in (threshold) voltage for our designed structure (Fig. 4) is 18.5 V, which is achieved from the ESSOLV simulation results. The displacement and stress distribution of this simulation are shown in Figs. 6 and 7, respectively. As it is evident in Fig. 6, the maximum stress on the membrane

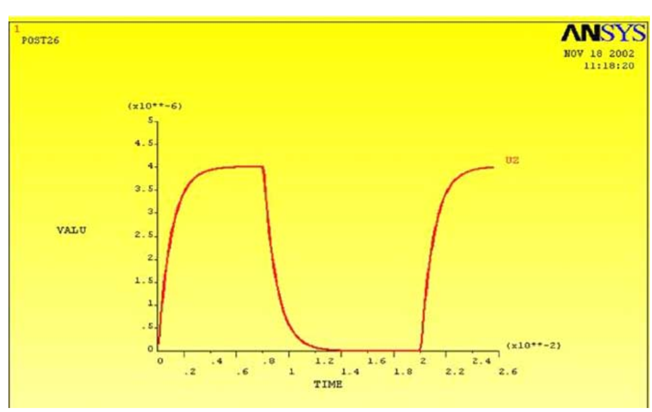

Fig. 8. Frequency response of the first membrane (x-axis: time (pumping cycle) (s); y-axis: membrane displacement (m)).

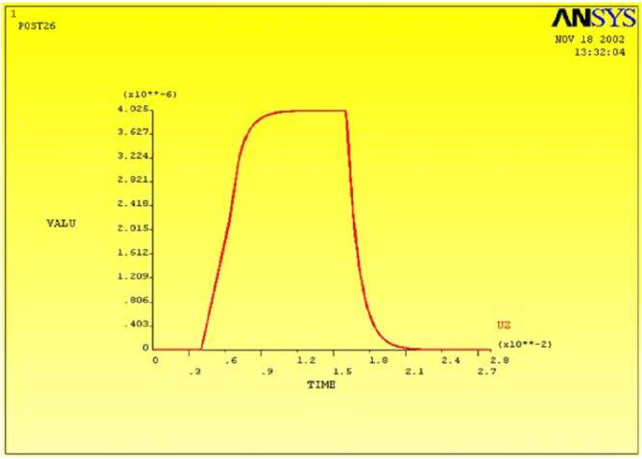

Fig. 9. Frequency response of the middle membrane (x-axis: time (pumping cycle) (s); y-axis: membrane displacement (m)).

is 31.7 MPa, which is much smaller than the critical stress of the polysilicon.

To calculate the flow rate of the designed micropump, we must consider the stroke volume of the first membrane. Therefore, we need the exact shape of the membrane deflection. Since, we do not have any analytical equation for our designed membrane deflection; finite element simulation is necessary.

The general equation for calculating the stroke volume is given by:

$$\Delta V = \int_0^a \int_0^a \int_0^H U(x, y)\mathrm{d}x\mathrm{d}y\mathrm{d}z$$

$$= \int_0^{1e-3} \int_0^{1e-3} \int_0^{4e-6} U(x, y)\mathrm{d}x\mathrm{d}y\mathrm{d}z \qquad (10)$$

where $U(x, y)$ gives the deflection of the membrane, $a = 1\mathrm{E}{-3}$ is the membrane length and $H = 4\mathrm{E}{-6}$ is height of the air gap. Since, $U(x, y)$ is analytically unknown for our structure, the only way to have an idea about stroke volume is to solve this equation employing finite element method. This method had been employed and the stroke volume is determined to be 3.0326 nl. Therefore, the flow rate is:

$$\text{Flow rate}: 3.0326\,\text{nl} \times 50\,\text{Hz} \times 60\,\text{s} = 9.1\,\mu\text{l/min}$$

This flow rate is true for the applied threshold voltage of 18.5 V. In practice to reach the desired volume flow rate, usually a calibration based on the variation of the pumping frequency or applied voltage is necessary.

To evaluate dynamic (frequency) response of the membranes, a pulsed voltage with frequency of 50 Hz is applied and its deflection behavior is considered. These simulations are shown in Figs. 8 and 9 for first and middle membranes, respectively.

6. Proposed fabrication process

Fabrication process of the designed micropump includes two substrates: Si substrate on which membrane part is constructed and glass substrate which includes input and output ports. These substrates are finally bonded together to form full structure. The process starts with a double side polished ⟨1 1 0⟩ oriented Si substrate. The step by step fabrication pro-

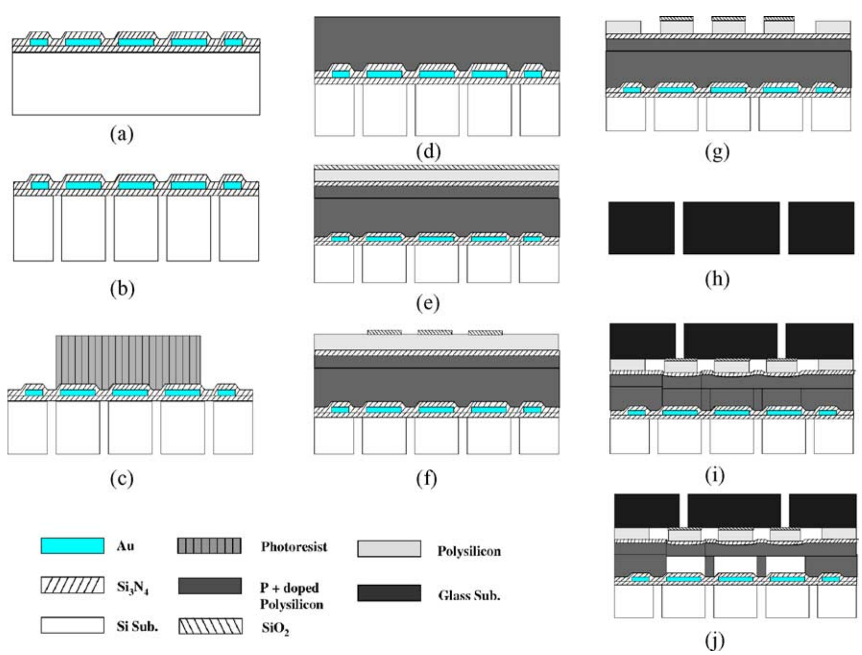

Fig. 10. Step by step proposed fabrication process of designed micropump (Au (▢); photoresist (▥); polysilicon (▢); Si_3N_4 (▨); p+ doped polysilicon (■); glass substrate (■); Si substrate (▢); SiO_2 (▧)).

cess for our structure is proposed as below and is shown in Fig. 10:

(a) A thin Si_3N_4 is deposited on the Si substrate which act as electrical insulation between pads and substrate.
(b) Au is sputtered and then patterned to form electrical pads.
(c) A thin Si_3N_4 is deposited and patterned. It plays the dielectric role between upper and bottom electrodes (Fig. 10a).
(d) Some holes are created on the backside of the Si substrate. These holes will be used to remove sacrificial layer (Fig. 10b).
(e) A 4-μm thick photoresist (sacrificial layer) is deposited and patterned. Since, the next processes will be done at the almost 200 °C and can affect the photoresist, it is cured in order to resist against high temperature processes (Fig. 10c).
(f) A 4-μm thick p+ doped polysilicon is deposited, planarized, and polished (Fig. 10d).
(g) The 2-μm/1000 Å/3-μm/0.5-μm p+ poly/Si_3N_4/poly/SiO_2 sandwich layer is deposited. The 2-μm polysilicon is the membrane thickness and the 3 μm is active valve thickness. SiO_2 is used as the selective anodic bonding layer and Si_3N_4 is used for etch-stop of the polysilicon (Fig. 10e).
(h) The SiO_2 is patterned to be used as the selective anodic bonding layer (Fig. 10f).
(i) The polysilicon is patterned. The Si_3N_4 is used as the mask (Fig. 10g).
(j) Two holes are created in the glass substrate which are the input and output ports (Fig. 10h).
(k) Two substrates are anodically bonded together (Fig. 10i).
(l) The photoresist (sacrificial layer) is removed. The Au and glass and Si_3N_4 act as the self-masking capability against the photoresist solution (Fig. 10j).

7. Conclusion

An electrostatic micromachined pump for medical applications is designed and simulated.

The micropump structure is based on the peristaltic mechanism, which has quite suitable pumping mechanism for drug delivery systems. Electrostatic actuation method is employed for the device. The size of our designed micropump is smaller than the other peristaltic micropumps especially piezoelectric counterpart. The size of micropump is 7 mm × 4 mm × 1 mm.

Active membrane valves are employed for flow control. Therefore, some problems associated with the passive check valves, such as valve clogging and valve breaking, are resolved. The simulated result for the threshold voltage of the micropump is 18.5 V. To achieve the variable pumping frequency and reach the desired flow rate, the actuation voltage is selected 1.2–1.5 times the threshold voltage, that is, 23 V for our design. The pumping frequency is considered 50 Hz. The proposed structure is simulated, using finite element methods by the ANSYS software and the results are in a good agreement with the drug delivery requirements.

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References

[1] J.W. Judy, Biomedical applications of MEMS, in: Measurement and Science Technology Conference, Anaheim, CA, 2000, pp. 403–414.
[2] N. Maluf, D.A. Gee, K.E. Petersen, Gregory.T.A., Kovacs Medical Applications of MEM, 2000, pp. 300–306, ISBN: 0780326369.
[3] J.G. Smits, Piezoelectric micropump with three valves working peristaltically, Sens. Actuators A21–A23 (1990) 203–206.
[4] L. Cao, S. Mantell, D. Polla, Implantable medical drug delivery systems using microelectromechanical systems technology, in: 1st Annual International IEEE–EMBS Special Topic Conference on Microtechnologies in Medicine and Biology, 12–14 October, Lyon, France, 2000, pp. 487–490.
[5] D. Maillefer, H. van Lintel, G. Rey-Mermet, R. Hirschi, A high-performance silicon micropump for an implantable drug delivery system, in: Proceedings of the MEMS ’99, January 17–21, Orlando, USA, 1991, pp. 541–546.
[6] T. Bourouina, A. Bosseboeuf, J.-P. Grandchamp, Design and simulation of an electrostatic micropump for drug-delivery applications, J. Micromech. Microeng. 7 (1997) 186–188.
[7] G. Lins, L. Skogberg, An Investigation of Insulin Pump Therapy and Evaluation of Using a Micropump in a Future Insulin Pump, M.S. Thesis, KTH, Stockholm, Sweden, 2001.
[8] P. Woias, Micropumps-summarizing the first two decades, in: C.H. Mastrangelo, H. Becker (Eds.), Microfluidics and BioMEMS, 2001, pp. 39–52.
[9] F.C.M. Van de Pol, H.T.G. Van Lintel, M. Elvenspoek, J.H.J. Fluitman, A thermopneumatic micropump based on microengineering techniques, Sens. Actuators A21–A23 (1990) 198–202.
[10] Q. Gong, Z. Zhou, Y. Yang, X. Wang, Design, optimization and simulation on microelectromagnetic pump, Sens. Actuators A83 (2000) 200–207.
[11] R.S. Shawgo, A.C. Richards, G. Yawen Li, M.J. Cima, BioMEMS for drug delivery, Curr. Opin. Solid State Mater. Sci. J. (2002) 329–334.
[12] L. Cao, S. Mantell, D. Polla, Design and simulation of an implantable medical drug delivery system using microelectromechanical systems technology, Sens. Actuators A94 (2001) 117–125.
[13] M.M. Teymoori, Design and Simulation of a Novel Electrostatic Micromachined pump for Medical Applications, M.S. Thesis, Department of Electrical Engineering, Urmia University, Iran, 2003.
[14] O. Français, I. Dufour, Normalized Abacus for the global behavior of membranes: pneumatic, electrostatic, piezoelectric or electromagnetic actuation, J. Model. Simul. Microsys. 1 (2) (2000).
[15] S.P. Timoshenko, S. Woinowsky-Kriefer, Theory of Plates and Shells, second ed., 1959, ISBN: 007Y858209.
[16] J.J. Yao, M.F. Chang, A surface micromachined miniature switch for telecommunications applications with signal frequencies from Dc up to 4 GHz, in: 8th International Solid-State Sensors and Actuators Eurosensors Conference, Stockholm, Sweden, June 25–29, 1995, pp. 384–387.

[17] P.J. French, Polysilicon: a versatile material for microsystems, Sens. Actuators A99 (2002) 3–12.

Biographies

Mir Majid Teymoori born in Urmia, Iran in 1979. He received his BSc degree in electrical engineering in 2000 and MSc degree in microelectronics (Micromachining) in January 2003, both from Urmia University. His current interests are design, simulation, and packaging of micromachined devices, such as micropumps, microswitches, and pressure sensors.

Ebrahim Abbaspour-Sani received a BSc degree from Ahwaz University, Iran in 1976, MSc degree from University of Wales Institute of Technology (UWIST), Cardiff, UK, in 1981 and PhD from University of New South Wales, Sydney Australia, in 1996. He joined the Department of Electrical Engineering, Urmia University, Iran, in 1983 and worked as a lecturer until 1992. He continued his academic activities with the Urmia University as an assistant professor after completion of his PhD degree in 1996. He is currently project director of the MEMS Laboratory at the Urmia University. His current interests are micro-machining technology applications on accelerometers, RF switches, and micropumps.