Computer simulation of electrostatic micropump

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An electrostatic micromachined pump is designed and simulated. The designed micropump has the advantages of flow rate controllability, self-priming, small chip size, and low power consumption. The designed micropump is simulated by the Runge-Kutta method. The flow rate of the designed micropump is considered in the range from10 to 50 µl/min which is quite suitable for drug delivery applications, such as chemotherapy. The predicted results for the first membrane deflection with different materials and at pulsed applied voltage is introduced.

في هذا البحث تم محاكاة تصميم لمضخة الكتروستاتيكية متناهية الصغر. من مميزات هذا التصميم التحكم في معدل الانسياب وصغر حجم الشريحة المستخدمة لتنفيذ هذا التصميم مما ينتج عنه فقد أقل في الطاقة. لقد تم استخدام طريقة رانج كوتا لبرنامج المحاكاة لتصميم المضخة. من الحسابات الخاصة بهذا التصميم وجد أن معدل الانسياب يتراوح بين ١٠ الى ٥٠ ميكرولتر لكل دقيقة مما يظهر أنه ملائم للتطبيقات الطبية لتوزيع العقاقير في المعالجة بالمواد الكيماوية. تم تطبيق برنامج المحاكاة لحساب انحراف الغشاء للمضخة مع مواد مختلفة.

Keywords: Simulation, Micropump, Electrostatic

1. Introduction

Microfluidics is a technology which refers to the research and development of micro-scale devices which handle small volumes of fluids (as small as micro-, nano-, pico and even femtolitre volumes). The devices themselves have dimensions ranging from several millimeters to micrometers and at least one of the dimensions of the device is often measured in micrometers, e.g., a channel in the device. Microfluidic devices require construction and design which differ from macro-scale devices [1]. 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, consumer electronics, communication, medical, nuclear, etc.

Microfluidic systems have diverse and widespread applications. Some examples of devices and systems include inkjet printers, portable blood analyzers, DNA and proteomic chips, lab-on-a-chip systems and micro total analysis systems. Applications can be found not only in diagnostics, pharmaceutics, biotechnology and environmental technology but in consumer electronics, pulp and paper and chemical, automotive and food industries as well. A number of medical devices and systems, such as blood pressure sensors, microneedles, glucose sensor, DNA analyzing system, etc., are designed and fabricated [2,3]. 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 [4].

Mechanical micropumps can be categorized according to the principles by which mechanical energy is applied to the fluid. In displacement pumps, such as peristaltic, reciprocating and rotary pumps, energy is periodically added by the application of force to a movable boundary. In dynamic pumps, such as ultrasonic pumps and centrifugal pumps, energy is continuously added to increase the fluid velocities within the pump. Non-mechanical pumps add momentum to the fluid by directly converting non-mechanical energy into kinetic energy without the mechanical movement of a structure. The driving forces can be electric, magnetic, thermal, chemical or surface tension forces [5].

A drug delivery device (micropump) must not introduce any toxic particles into the drug and vice versa [6]. 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 [7, 8], 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 [9]. It should be noted that the flow rate of medical micropumps must be controlled at all times.

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

2. The micropump structure design

The structure of the electrostatic micropump is shown in fig. 1. It consists of three layers: glass, Si substrate, and membrane part [10]. Membrane part includes three active valves on top of the membranes, microchannels, three electrostatic chamber (air gap), input, and output. Working principle is based on the peristaltic motion which is schematically shown in fig. 2. Dead volume of this proposed structure is very small and the compression ratio is almost 0.8 which guarantees the self-priming capability of this micropump.

3. Simulation of the electrostatic micropump

Simulation of micromachined systems is becoming increasingly important. The motivation here is similar to that of the simulation of purely electronic VLSI circuits: before fabricating a prototype, one wishes to virtually build the device and predict its behavior. This allows for the optimization of the various design parameters according to the specifications. As it is a virtual device, parameters can be changed much more quickly than actually fabricating a prototype, then redesigning and fabricating it again. This considerably reduces the time to market and also the cost to develop a commercial device. Simulation software tools for electronic circuits are very mature nowadays, and the level of realism is striking. Often the first fabricated prototype of a novel circuit works in a very similar way as predicted by the simulation. In MEMS, however, this degree of realism cannot be achieved in many cases for two reasons. First, the simulation tools have not reached a similar maturity as their electronic equivalents; and second, and more importantly, simulation of MEMS devices is much more complex. A MEMS device typically comprises many physical domains such as mechanical, electrical, thermal, and optical. All these domains interact and influence each other, making the problem orders of magnitude more difficult [11].

Many of the traditional software solutions have add-ons for models and simulations of microfluidic devices and phenomena. They are based on the application of various constraints on the model. Some of them can be combined with other phenomena and thereby provide metaphysics modeling. These tools allow for modeling situations which would otherwise be difficult to test in reality or can be used for the fine-tuning of the structure before its actual prototyping, in order to avoid high cost.

The governing mechanical equation for a completely all sides-fixed membrane with uniform thickness (fig. 3) is given by [12]:

$$Dh^3\Delta g(x,y) - Th\Delta g(x,y) = p(x,y)\,, \qquad (1)$$

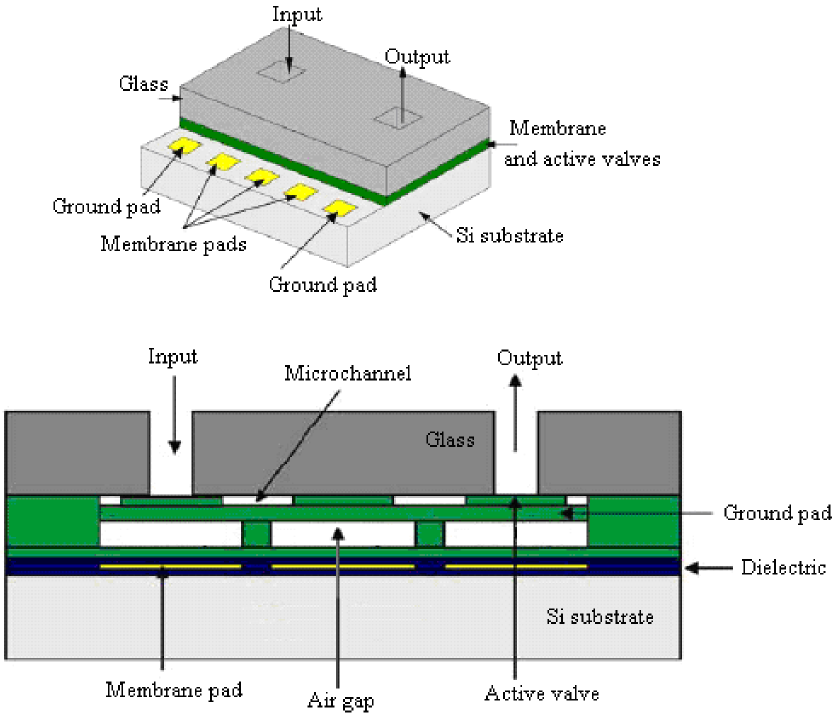

Fig. 1. The schematic of electrostatic micropump.

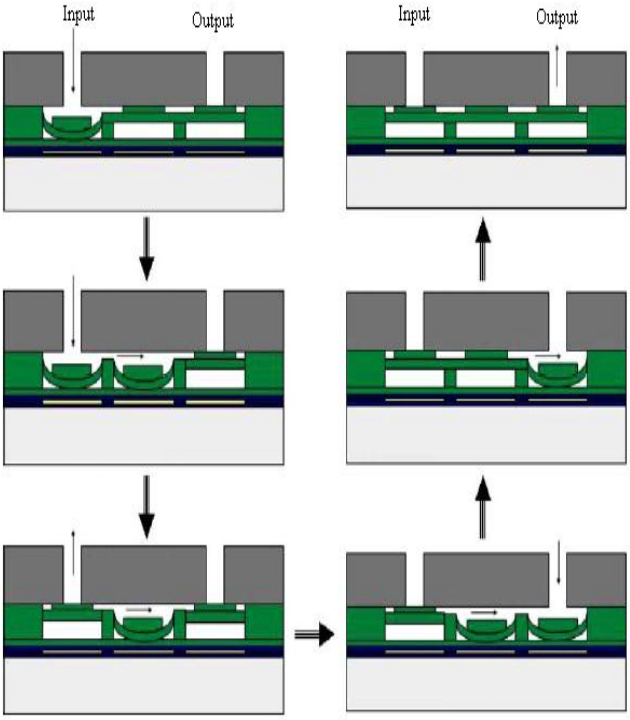

Fig. 2. Working principle of the micropump.

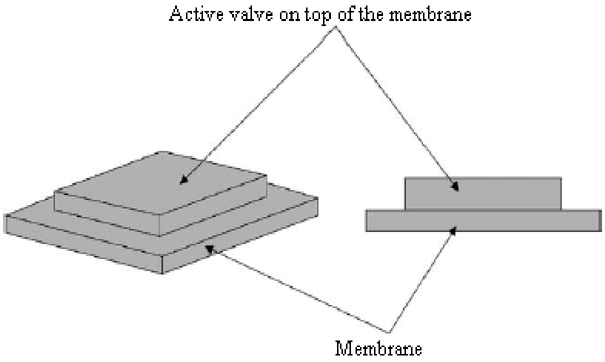

Fig. 3. Active membrane valve.

where $\Delta = \partial^2/\partial x^2 + \partial^2/\partial y^2$, T is the residual stress on the membrane, h is the membrane thickness, and D is the mechanical rigidity:

$$D = E/12(1-\upsilon^2)\,, \tag{2}$$

$$p(x,y) = \frac{\left(\varepsilon V^2/2g_o^2\right)}{\left(1-\left(\omega(x,y)/g_o\right)\right)^2}\,, \tag{3}$$

where: g_o is the gap between two electrodes (μm), ε is the permittivity, $\omega(x, y)$ is the deflection of the membrane (μm), V voltage applied E Young's modulus (Pa), and ν Poisson's ratio..

The governing dynamic equation for a completely fixed membrane with voltage V applied between the membrane and the fixed electrode is:

$$m\frac{\partial^2 Z}{\partial t^2} + b\frac{\partial Z}{\partial t} + KZ = \frac{\varepsilon_o A V_s^2}{2g_o^2}\,, \tag{4}$$

with initial conditions of: at start (t=0):
The dynamic response of the membrane (Z) = 0, and $\partial Z/\partial t = 0$.

Where m = effective mass of the membrane, b= damping coefficient, V_s = the applied voltage, A the plate area, Z is the displacement, and K is the spring constant.

Assuming no damping (b=0) then:

$$Z(t) = \frac{\varepsilon_o A V_s^2}{2Kg_o^2}(1-\cos(\sqrt{\frac{k}{m}}t))\,. \tag{5}$$

The spring constant is given by eq. (6):

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

Where a, h membrane dimension (a length – h thickness).

The threshold voltage is given by:

$$V_{th} = \sqrt{\frac{8Kg_o^3}{27\varepsilon A}}\,. \tag{7}$$

We consider A =1.5 mm x 1.5 mm, g_o= 4 μm, ε=8.854E -12

In the present study a is assumed 1500 μm, h= 3μm.

4. Numerical simulation results

In order to research the micropumps dynamics, it is necessary to create mathematical models according to the fluid transmission procedure and the element characteristics [13], table 1.

The Runge-Kutta method [14] can be applied to solve the above equations. The analytical results for the first membrane deflection with different materials and at pulsed applied voltage with frequency of 50 Hz is shown in fig. 4.

Table 1
The micropump structure characteristics

	Glass	SiO_2	Au	Si_3N_4
Young's modulus (G Pa)	41.6-62.4	70	80	195.05-315.05
Poisson's ratio	0.166	0.17	0.44	0.24
Density (kg/m^3)	2250	2200	19280	3187

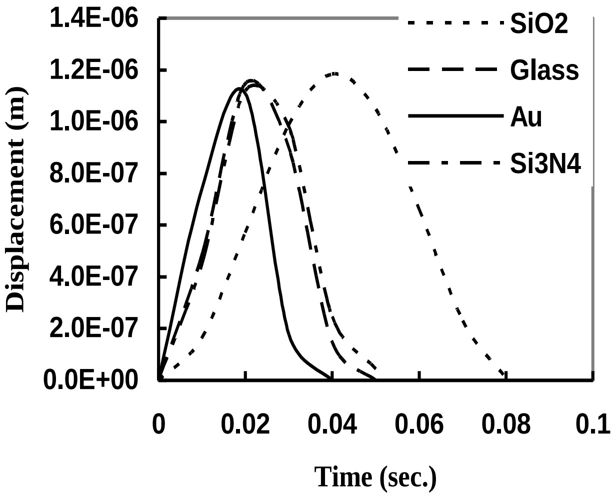

Fig. 4. The membrane deflection for various materials.

The membrane length is taking as a=1E-3 and the height of the air gap is assumed 4E-6 the volume flow rate is given by:

$$Q = \int_0^{1e-3} \int_0^{1e-3} \int_0^{5e-6} u dx dy dz * f \,, \qquad (8)$$

where u is the membrane deflection and f is frequency. This equation can be calculated numerically to find the value of the flow rate that is range between 10 to 50 µl/min in this study depending on the membrane material used.

The goal of this analysis and simulation is to determine the micropump dimensions and use the mechanical analysis and simulation results to direct micropump fabrication and reduce fabrication cost.

5. Conclusions

An electrostatic micropump based on the peristaltic mechanism, which has quite suitable pumping mechanism for drug delivery systems is simulated. The designed micropump has the advantages of flow rate controllability, self-priming, small chip size, and low power consumption. The flow rate of the designed micropump is considered in the range from 10 to 50 µl/min which is quite suitable for drug delivery applications, such as chemotherapy. The predicted results for the first membrane deflection with different materials is introduced.

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