Design and Simulation of Valveless Piezoelectric Micropump

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Abstract

In this paper some discrete parts of a valveless piezoelectric micropump for drug delivery system is designed and simulated. The core components of the micropump are actuator unit that converts the reciprocating movement of a diaphragm actuated by a piezoelectric actuator into a pumping effect and Nozzle/diffuser elements that are used to direct the flow from inlet to outlet. Simulations are performed for actuator unit and diffuser/nozzle element individually, using Comsol software. The simulation results show that displacement of PZT actuator is directly proportional to the applied electric field. Flow is greater at contraction and lower at expansion for the diffuser/nozzle elements.

Keywords: Valveless Micropump, Diffuser/nozzle, piezoelectric actuator.

1. Introduction

A Microelectromechanical system is a rapidly growing field which enables the manufacture of small devices using microfabrication techniques similar to the ones that are used to create integrated circuits. MEMS technologies have been applied to the needs of biomedical industry giving rise to a new emerging field called Microfluidics. Microfluidics deals with design and development of miniature devices which can sense, pump, mix, monitor and control small volumes of fluids. The development of microfluidic systems has rapidly expanded to a wide variety of fields. Principal applications of microfluidic systems are for biological and chemical sensing, drug delivery system (DDS), lab-on-a-chip (LOC), point of care testing (POCT), molecular separation such as DNA analysis, amplification and for environmental monitoring.

Conventional drug delivery methods such as oral medications, inhalers and subcutaneous injections do not deliver all drugs accurately and efficiently within their desired therapeutic range. Generally most of the drugs are effective if delivered within a specific range of concentration between the desired maximum and minimum levels. Above the maximum range, they are toxic and below the minimum range, they have no therapeutic benefit. With controlled drug delivery systems consisting of drug reservoir, micropumps, valves, microsensors, microchannels and necessary related circuits, appropriate and effective amount of drug can be precisely calculated by the controller and released at appropriate time by the microactuator mechanism such as micropump [1]. Micropumps are therefore an essential component in the fluid transport systems.

2. Related Work

Research on micropumps was initiated in 1980 and numerous different pumps have since been developed. The pumps that have attracted most attention are piezoelectric pumps, mainly because of the broad range of fluids which can be pumped and because the pumps are readily realized using silicon micromechanics. In order to provide precise and repeatable pumped volumes of fluids with each cycle and to reduce the valve leakage problem peristaltic

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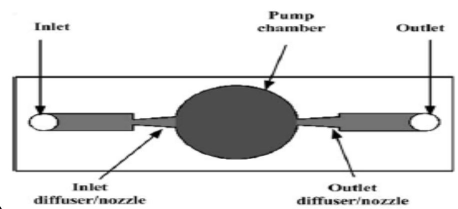
micropump was designed [6]. A micropump with diffuser/nozzle elements as flow rectification elements instead of valves was proposed. This eliminates check valve fatigue and valve clogging. The design features includes deep diffusers, a shallow pump chamber, and a thin pump diaphragm. [7]. The literature survey indicates the latest research being focused on improving the efficiency, flow rate and reducing the back flow pressure of micropump through physical and geometrical configuration. It also indicates that all the micropumps were designed and simulated using water as a working fluid. Accordingly, in the present work, a valveless piezoelectric micropump with diffuser/nozzle elements for flow rectification and based on reciprocating pumping principle for Gentamicin intravenous administration is studied for MEMS applications. In the design of valveless diffuser/nozzle micropump, deep pump chamber is proposed instead of shallow pump chamber in order to increase the flow rate. The model is designed and simulated using 'Comsol Multiphysics' software package. The analysis concentrates on improving the efficiency, flow rate and reducing the back flow pressure of micropump.

3. Design of the micropump

A valve less piezoelectric micro pump basically includes two diffuser/nozzle elements, pump chamber, actuator unit, inlet and outlet channel, power supply module and diaphragm/pump membrane as shown in Fig 1.

Actuator is necessary to operate the diaphragm of the micro pump. The actuator is made of a piezoelectric disc and a silicon membrane. The piezoelectric disc is made up piezoelectric material such as Lead Zirconate Titanate (PZT-5A) to yield mechanical strain by an external electric field. The diffuser/nozzle determines the performance of the micropump. Two flat walled diffuser made up of silicon is selected for the proposed micropump design, which has a rectangular cross section with two parallel flat walls and two convergent flat walls as shown in Fig. 2. The basic dimensions of the diffuser element involve the divergence angle θ, the diffuser length L and the width of the narrowest part W_1 [2].



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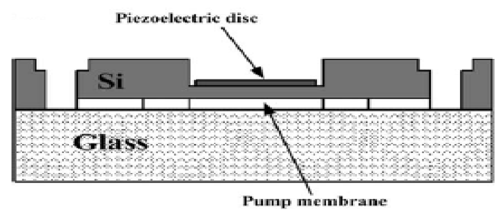

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Fig.1 Schematic of valve less micro pump (a) top view and (b) side view

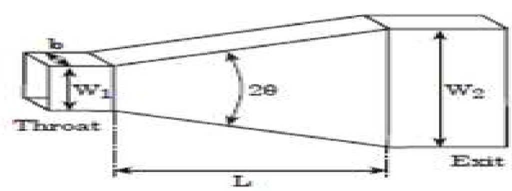


Fig. 2 Flat-walled diffusers

One inlet and one outlet flow hole is drilled in the inlet/outlet cavity of the micropump. Two short brass pipes are fixed to the pump inlet and outlet holes for the tube attachment. The pump chamber having circular cavity made up of silicon and covered by glass substrate is considered. The diaphragm closes the cavity of the pump chamber and it is bonded with the centre disk of the actuator. In reciprocating pumping method, flow rate of micropump depends on the stroke volume, which is the volume change of chamber caused by diaphragm motion. Circular diaphragm of 0.1mm thickness and 6mm in diameter of silicon material is considered. An alternating voltage of 50V (0 to peak) is applied across piezoelectric disc. This voltage is required to be switched in the form of on and off such that required frequency of this voltage can be applied to the PZT stack to achieve pumping action. As a result, a control unit is interfaced between power supply module and micro pump.

4. Principle of Operation of Micropump

The pump operation is based on the fluid flow rectifying properties of the two nozzle/diffuser elements. The dimension difference at both ends of the diffuser causes the pressure difference and drives the fluid. The pump cycle of the pump is divided into a ‘supply mode’ and a ‘pump mode’. In the supply mode, the actuator increases the chamber volume, resulting in lower pressure inside the chamber. As a result, a larger amount of fluid flows into the chamber through the input element, which acts as a diffuser, than through the input element, which acts as a nozzle. In the pump mode, the actuator decreases the chamber volume, resulting in higher pressure inside the chamber. As a result, a larger amount of fluid flows out of the chamber through the output element, which acts as a diffuser, than through the output element, which acts as a nozzle [2].

5. Analytical Model of Micropump

The analytic model of valveless piezoelectric micropump is shown in Fig. 3. The pressure in the pump chamber is P. P_{in} and P_{out} are the inlet and the outlet pressures, respectively. Q_{in} and Q_{out} are the volume flow through the inlet and the outlet, respectively.

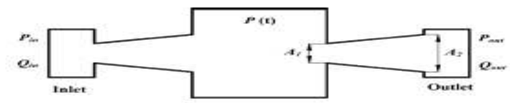


Fig. 3 Analytical model of a valveless piezoelectric micropump

At the resonance frequency, the cavity volume variation due to the oscillating diaphragm can be expressed as:

$$V_c = V_o \sin 2\pi f_o t \tag{1}$$

where $$V_o = K_v x_o \tag{2}$$

where V_0 is the volume variation amplitude, K_v is a constant, x_0 is the diaphragm center deflection amplitude, f_0 is the pump excitation frequency and t is the time [4]. The pressure drop across a diffuser and nozzle can now be written as:

$$\Delta p_d = \xi_d \frac{1}{2} \rho u_d^2 \tag{3}$$

and

$$\Delta p_n = \xi_n \frac{1}{2} \rho u_n^2 \tag{4}$$

Here ξd and ξ_n are pressure loss coefficient of the diffuser and nozzle, u_d and u_n are the fluid flow velocities in the narrowest parts of the diffuser and nozzle [3]. The flows Q_d and Q_n across the diffuser/nozzle directions are respectively expressed as:

$$Q_d = A_1 \left(\frac{2}{\rho}\right)^{1/2} \left(\frac{\Delta p_d}{\xi_d}\right)^{1/2} \tag{5}$$

$$Q_n = A_1 \left(\frac{2}{\rho}\right)^{1/2} \left(\frac{\Delta p_n}{\xi_n}\right)^{1/2} \tag{6}$$

Assuming the flow to be incompressible, the simplified governing equation of valveless micropump is obtained as:

$$\frac{\partial V_c}{\partial P}\frac{dP}{dt}+\frac{\partial V_c}{\partial t}=Q_{in}-Q_{out} \qquad (7)$$

6. Simulation and Results

The study of a valveless piezoelectric micropump results in coupled partial differential equations in electrical, mechanical and fluid-solid domain. These either cannot be solved analytically or lack an exact analytic solution due to the complexity of the boundary conditions or domain. The finite element method is used to solve the problem. The commercially available FEM-based software Comsol is used to simulate the valveless micropump. Simulations are performed for actuator unit and diffuser/nozzle element individually.

6.1 Simulation of actuator unit

The 3D builder module of Comsol is used to build the actuator unit. The actuator is made of a piezoelectric disc with the dimension of Φ6mm × 0.15mm thick and a silicon membrane with the dimension of Φ6mm× 0.1mm thick. The pump flow depends on the excitation frequency exerted on the piezoelectric actuator and the deflection shape of the pump membrane. The 10-node is chosen to model the actuator to perform Eigen frequency analysis to determine the vibration characteristics i.e. natural frequencies and natural mode shapes of a structure during free vibrations while it is designed. Figure 4 shows the results of Eigen frequency analysis. The first natural frequency is 58605.12Hz and the second natural frequency is 119671Hz. The piezoelectric actuator bends to one direction under the first natural frequency and it has only one peak. The piezoelectric actuator has two peaks under the second natural frequency.

Then Piezoelectric Actuator is excited at the working voltage of 50V (0 to peak) and the excitation frequency much lesser than the natural frequency. The maximum deflection obtained is 0.176μm at the centre of pump membrane.

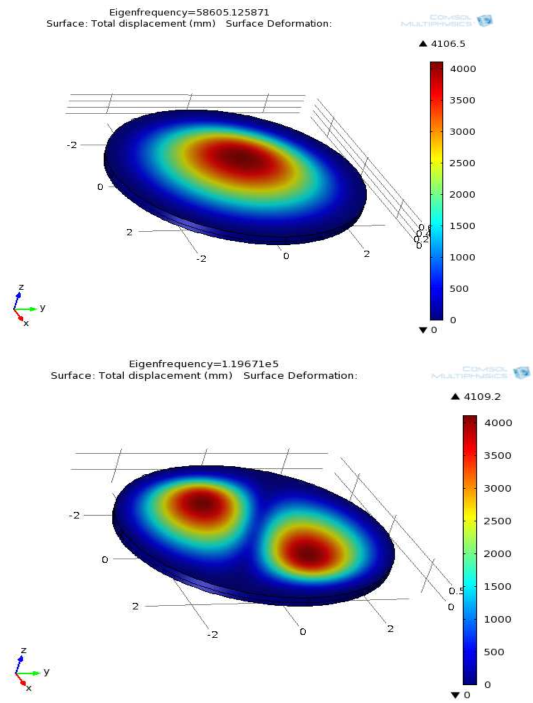


Fig 4. The modal shape of the piezoelectric actuator at the first and second natural frequencies

Figure 5 shows the relationship between the displacement at the central of pump membrane and excitation frequency for different excitation voltages. The excitation voltage is varied from 80V to 120V and the excitation frequency is varied from 200Hz to 500Hz. It is observed that the excitation voltage is a more important factor affecting the central displacement compared with the excitation frequency. The displacement of PZT actuator is directly proportional to the applied electric field.

The reason for considering the excitation frequency lesser than the natural frequency of piezoelectric actuator, because at this range of frequency both the reaction of fluid membrane coupling and the inertial force of membrane are very small relative to the elastic deformation force of membrane. Hence the nonlinear effect of the fluid-membrane coupling at zero pump pressure is negligible [5].

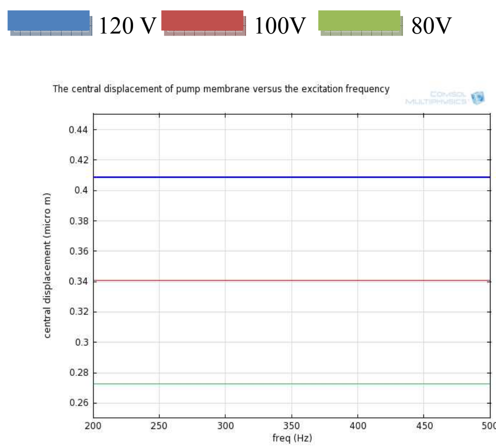


Figure 5. The central displacement of the pump membrane versus the excitation frequency

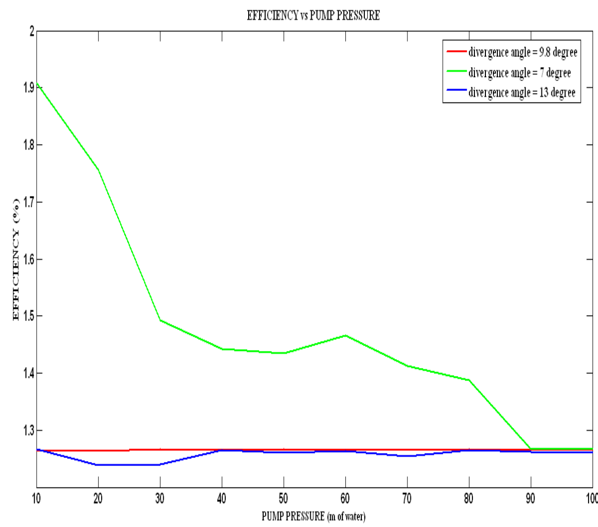
6.2 Simulation of diffuser element

The 2D builder module of *Comsol* is used to build the Diffuser element. Water is used as the working fluid, thereby limiting the problem to incompressible flow. The diffuser model is imported into *Laminar Flow module* to measure diffuser element efficiency ratio and volume flow rate of diffuser. Stationary analysis is performed for different diffuser elements with dimensions mentioned in Table 1, with the diverging-wall direction as positive direction.

Table 1: Dimensions of Diffuser elements

NOTATION	W1 [µm]	L [µm]	α = 2θ
3a	80	1093	9.80
3b	80	1440	9.80
3c	80	1093	7.00
3d	80	1093	130

Figure 6a shows that the smallest opening angle provides best flow directing capability. Figure 6b shows that the smallest opening angle provides best pump performance. At low pressure, diffuser element with divergence angle, $\alpha = 9.8^0$ shows better pump performance compared to the diffuser element with divergence angle, $\alpha = 13^0$. Figure 7a shows that the longer element provides best flow directing capability. Figure 7b shows that the shorter element provides best pump performance.



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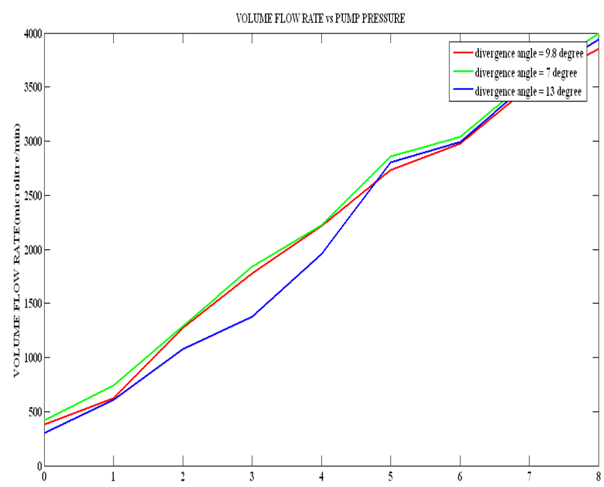
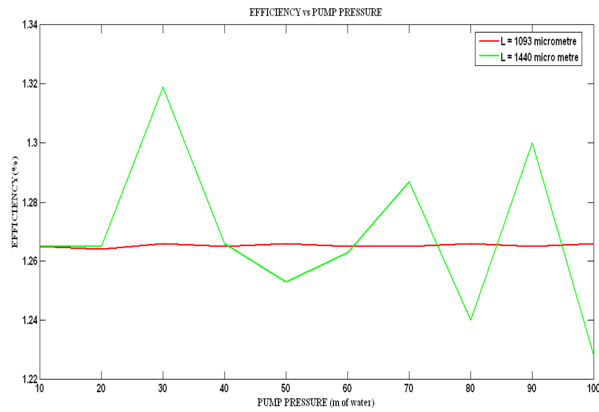


(b)

Fig.6 (a) Diffuser element efficiency ratio versus pump pressure for different divergence angles (b) Diffuser element volume Flow rate versus pump pressure for different divergence angles



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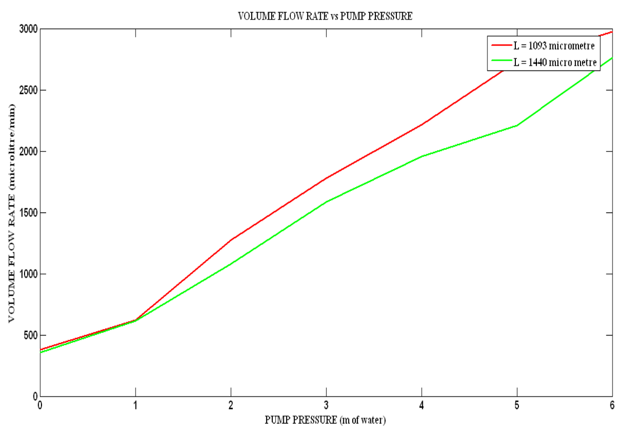
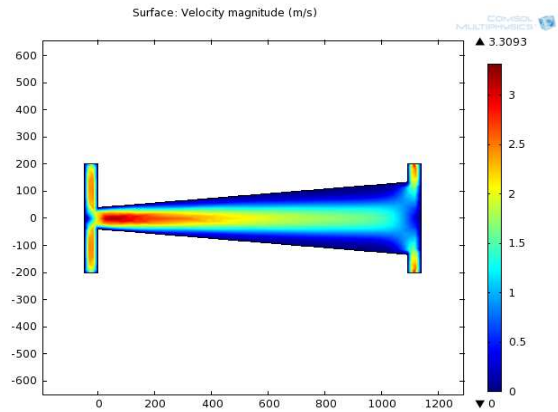


(b)

Fig. 7 (a) Diffuser element efficiency ratio versus pump pressure for different diffuser lengths (b) Diffuser element volume flow rate versus pump pressure for different diffuser lengths

The two dimensional model of a diffuser element of length 1093μm, opening angle of 9.8^0 and smallest width of 80μm with the diverging-wall direction as positive direction and the converging-wall direction as positive direction is simulated. Simulations are performed for the pressure range 0-100kPa, using water as a working fluid, thereby limiting the problem to incompressible flow. In diverging-wall direction, the velocity is reduced before the exit before the remaining kinetic energy is lost in a jet at the outlet as shown in Fig 8a. In the converging-wall direction, the flow is accelerated through the nozzle to high velocity and the kinetic energy is lost in a jet at the outlet as shown in Fig 8b.



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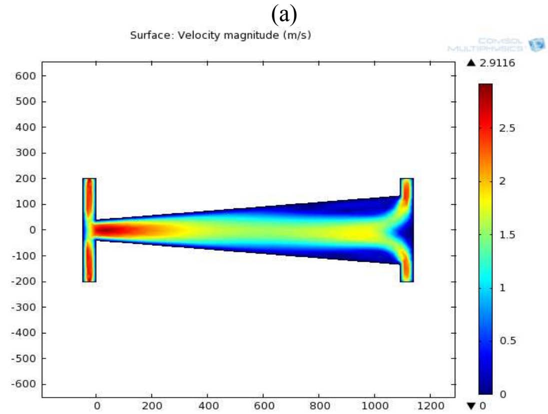

(b)

Fig. 8 (a) Velocity vector plot with entering pressure 13kPa and 0Pa backpressure at outlet (b) Velocity vector plot with entering pressure 13kPa and 0Pa backpressure at inlet

7. Proposed Work

Similar simulations are performed for diffuser element with same dimension, with the diverging-wall direction as positive direction using Gentamicin as a working fluid. The simulated diffuser element efficiency ratio versus pump pressure and volume flow rate versus pump pressure for diffuser elements with 80×80μm diffuser throat cross-section and divergence angle, $\alpha = 9.8^0$, and diffuser length, L = 1093μm for Gentamicin is shown in Fig 9.

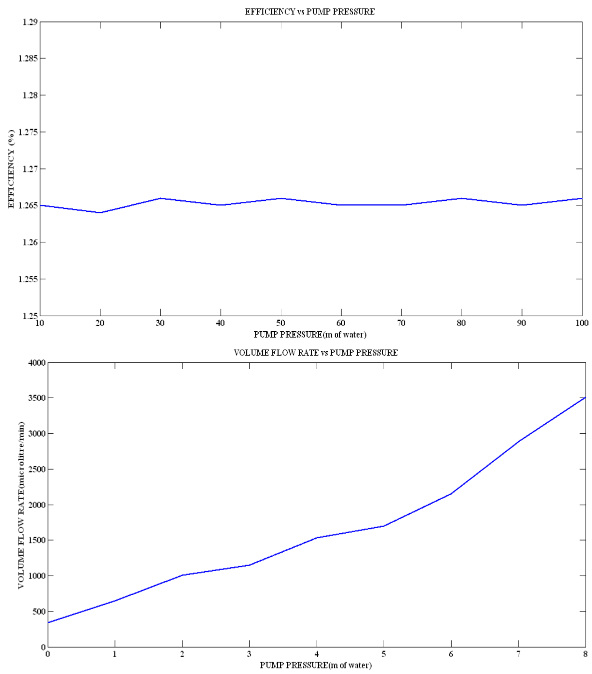

Fig. 9 Diffuser element efficiency ratio versus pump pressure and Diffuser element volume flow rate versus pump pressure

8. Conclusion

A micropump designed on the basis of PZT actuator for gentamicin intravenous administration is presented. The simulation was performed for 3D membrane using Comsol package and results conclude that deflection of actuator is linear with applied potential over PZT material of the membrane. The maximum deflection observed was 0.176µm at 50 volt 0-to-peak voltage. On the other hand a 2D diffuser/nozzle element was simulated for the laminar flow to check the rate of change of flow from expansion towards the contraction and vice versa. Flow is greater at contraction and lower at expansion for the diffuser/nozzle elements.

9. References

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