

Design, Simulation and Study of Micro-pump, Micro-valve and Micro-needle for Biomedical Applications

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Abstract: This paper addresses the various aspects of design and simulation of MEMS based micro-pump, micro-valve and micro-needles. Solid stress/ strain, piezoelectric, microfluidic and moving mesh application modes of the COMSOL MULTIPHYSICS Software package was used for the purpose of design and simulation of the models. The data obtained from the simulation results were used to calculate the operating frequency and flow rate of the piezoelectrically actuated micro-pump. The PDMS flap based micro-valve prevents back flow of fluid by operating like a microfluidic diode. Delivery of drugs and sampling of body fluids from patient's body can be performed using silicon micro-needles. These devices can be combined to develop an automatic drug delivery system that has potential use in the healthcare sector.

Keywords: Micro-pump, micro-valve, micro-needles

1. Introduction

Although medical science has made phenomenal advancements in prevention, control and cure of various diseases, this can be of limited use in remote areas where there is a scarcity of trained medical practitioners and technicians. In such situations automated and easy to use solutions are necessary that can allow sampling and testing of body fluids and precise introduction of medicine and nutrients in a patient's body. In recent years, the advent of new emerging interdisciplinary research areas like biomedical engineering, automatic drug delivery systems, lab-on-a-chip devices etc. have fueled the interest in research in such systems that are likely to improve the quality-of-life in remote and backward areas. In this paper we have designed, simulated and investigated three major components of such systems, the micro-pump, micro-valve and micro-needle.

Displacement-type micro-pumps are one of the most common types of mechanical micro-pumps

that consist of one or more pump chambers closed with flexible diaphragms, which are moved periodically by some actuating mechanism [1]. The periodic motions of the diaphragms cause periodic change in the fluid volume and pressure in the pump chambers. The flow rectifying elements, such as valves and diffuser/ nozzle elements are attached to the pump chambers and act as the inlet/ outlet ports directing the fluid from one direction to other. Arrays of micro-needles either in a lateral or a vertical geometry may be attached to the micro-pump which then forms a complete system capable of almost painless drug delivery. Backflow of the pumped fluid under the influence of back pressure is undesirable in certain situations, including drug delivery. The effect of back pressure can be diminished by introducing a microfluidic diode or micro-valve in the channel between the outlet of the pump and the needle array. The micro-pump, micro-valve and the micro-needles can be combined to form a complete drug delivery system capable operating independently without human intervention.

In this paper, we present our work involving micro-pumps, micro-valves and micro-needles designed and simulated using the COMSOL MULTIPHYSICS software package.

2. Design

2.1 Design of Micro-pump

The basic structure of the proposed micro-pump is made of two chambers and an inlet, connected to each other by diffuser-nozzle elements. PMMA membranes cover both the chambers and a piezoelectric material (PZT-5H) is deposited on the PMMA membrane atop the chambers. The schematic diagram [Fig.1] depicts the basic structure of the micro-pump. Each of the two pump chambers has a dimension of 5mm × 5mm × 0.2mm, which can be fabricated by bulk etching of silicon. An array of vertical micro-needles can be fabricated and on the other

side (bottom) of the chamber, which can be used for proposed drug delivery purpose.

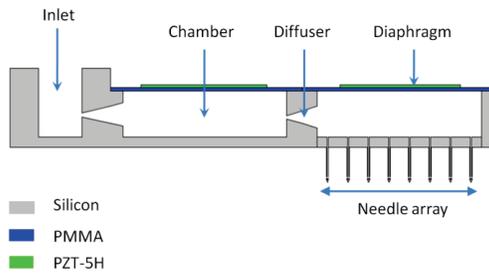


Figure 1. Structure of the Micro-pump

2.2 Design of Micro-valve

The micro-valve is basically a flap made of Poly-Di-Methyl-Siloxane (PDMS) which bends under the influence of fluid velocity and opens a channel in the flow direction and seals in the opposite direction. The device has a PDMS flap at the junction of two channels of different cross-sections [Figure 2]. The heights of channel 1 and channel 2 are $50\ \mu\text{m}$ and $100\ \mu\text{m}$ respectively. The height and width of the flap and its distance from the channel junction are varied in order to perform a detailed analysis about the flow rate of the device. The back pressure required to close the flap and its dependence on various flap parameters is also studied.

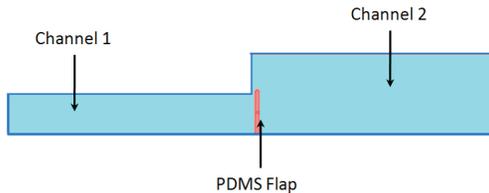
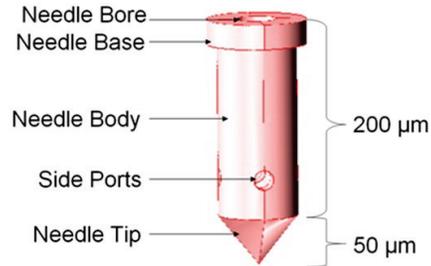


Figure 2. Structure of the Micro-valve

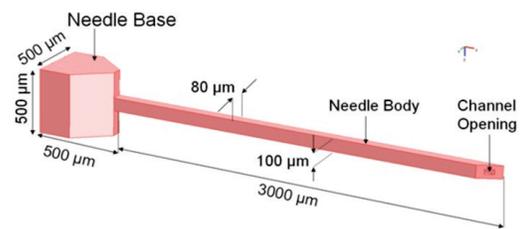
2.3 Design of Micro-needle

Different types of micro-needles are often required for different types of drug delivery applications. An out-of-plane silicon micro-needle (length $250\ \mu\text{m}$) can be used for insertion of fluid into the dermis layer of human skin [Figure 3(a)]. For the purpose of insertion of drugs and extraction of body fluid from the subcutaneous fat layer an in-plane needle (length $3000\ \mu\text{m}$) [Figure 3(b)], can be used. Both types of micro-needles have sharp wedge like tip angles to minimize skin resistance force and

insertion pain. The needle channels have been opened through side ports at the tip to prevent clogging of the channel by skin debris during insertion of the needle.



(a) Out-of-plane Micro needle



(b) In-plane Micro needle

Figure 3(a) and (b). Two types of Micro-needles

3. Use of COMSOL Multiphysics

The models of micro-pump, micro-valve and micro-needles were designed and simulated using COMSOL Multiphysics. The application modes used for the simulation of the micro-pump are piezoelectric, solid stress/ strain and microfluidic modules. For the purpose of microfluidic simulation, only the internal fluid part of the pump was modeled, excluding the solid parts.

The application modes used for the simulation of micro-valve are solid stress/ strain, microfluidic (incompressible Navier-Stokes), and moving mesh (ALE). One of the most important characteristics of the micro-valve model is the implementation of fluid-structure interaction.

The structural simulations of the silicon micro-needles were performed using the

structural mechanics solid stress/ strain module. For the purpose of microfluidic simulation, the internal channels of the micro-needles were modeled using the incompressible Navier-Stokes module.

4. Simulation Results and Discussions

4.1 Simulation of the Micro-pump

The piezoelectrically actuated micro-pumps are made of a pressure chamber closed by a flexible diaphragm and flow directing elements such as diffuser and nozzle. On condition that the diffuser/nozzle elements have been designed properly, the fluid flow in the diffuser direction would be higher than the fluid flow in the nozzle direction, assuming the same pressure drop across the fluid element. The pump cycle of the micro-pump can be divided into a 'supply mode' and a 'pump mode' [Figure 4 (a) and (b)].

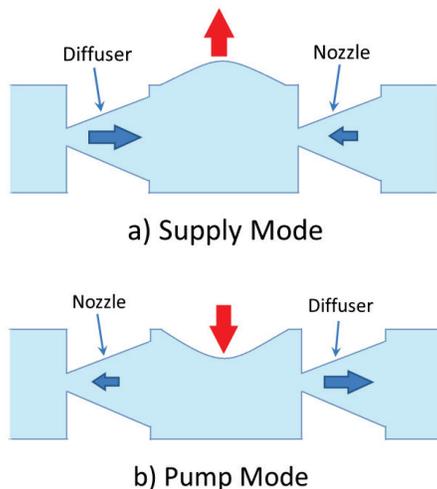


Figure 4. (a) Supply and (b) Pump mode

During the 'supply mode' the pump diaphragm deflects upward under the influence of upward bending of the piezoelectric material. Consequently, the cavity volume increases so that a larger amount of fluid flows into the cavity through the input element, which acts as a diffuser, than through the output element, which acts as nozzle. On the contrary, during the 'pump mode' the diaphragm deflects downward due to downward bending of the piezo material. This results in a decrease of cavity volume, allowing a larger amount of fluid to flow out of the cavity through the output element, which acts as a

diffuser, than through the input element, which acts as a nozzle. The complete pump cycle of the piezoelectrically actuated diffuser/ nozzle micro-pump drives a net volume of fluid from the input to the output side of the pump. In order to achieve the best pump performance, the excitation frequency should be close to the pump resonance frequency. The resonance frequency of the micro-pump is governed by the mass of the fluid in the diffuser/ nozzle elements, the properties of the fluid and the elastic properties of the pump diaphragm.

The simulation of the solid model of the pump, consisting of silicon body and needles and a diaphragm made of PMMA and PZT-5H was performed in COMSOL MULTIPHYSICS piezoelectric stress/ strain mode. The chosen boundary condition was that the silicon body of the pump remain fixed, while electric potentials were applied on the top of the PZT-5H material.

The variations of the deflection of the diaphragm for different applied voltages were studied. It was observed that the diaphragm deflection has an almost linear dependence on the applied voltage [Figure 5]. Pumping action is achieved when the pump diaphragms are operated in out-of-phase manner. In order to achieve the out-of-phase operation of the two diaphragms, two sinusoidal voltages of equal amplitude and frequency are applied on the two PZT-5H actuators attached to the pump diaphragms at a phase difference of 180°.

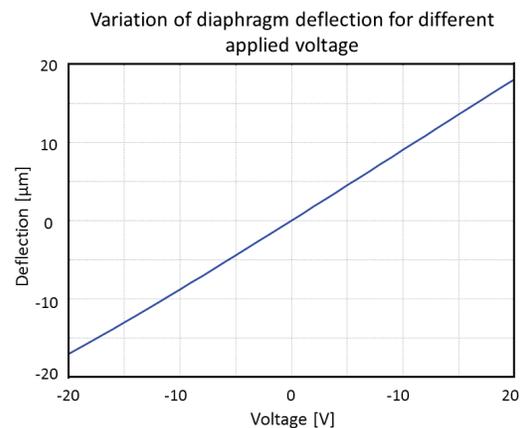


Figure 5. Variation of diaphragm deflection with different applied voltages

In response to the periodic deflection of the pump diaphragms, the fluid pressure and

velocity in both the chambers also change periodically, as shown in Figure 6. Although periodic in nature, the variation in pressure and velocity is not sinusoidal, since the fluid mass flowing through the nozzle and the needles tend to have an effect on the fluid velocity in the pump chambers.

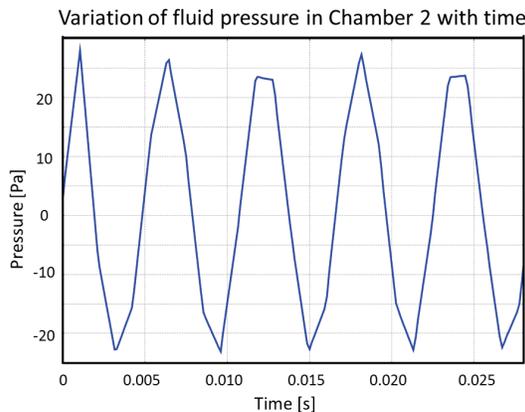


Figure 6. Variation of maximum fluid pressure in Chamber 2

Figure 7 shows the variation of fluid flow rates through the inlet (diffuser) and outlet (needle array) of chamber 2.

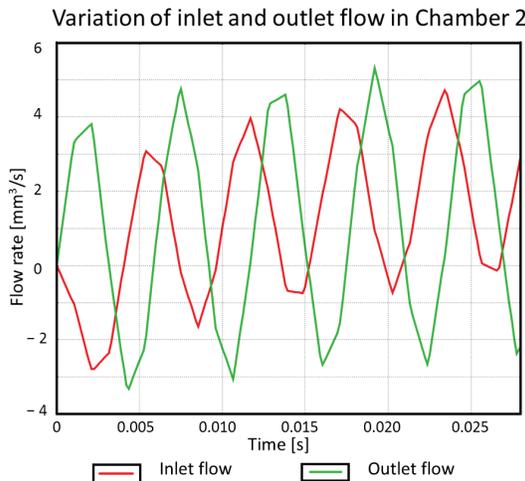


Figure 7. Variation of inlet and outlet flow rates in Chamber 2

From Figure 7 it can be noted that the total volume flow rate, which can be obtained by performing time integral of the flow rates through the needle array, is positive, indicating a net flow of fluid from the inlet to the needle array.

Taking into consideration all the related parameters such as diffuser/nozzle dimensions, properties of the diaphragm and the pumped fluid (water) etc. the pump resonance frequency is found to be 169.8 Hz. At this operating frequency the pump flow rate is calculated as 375 $\mu\text{l}/\text{min}$ [3].

4.2 Simulation of the Micro-valve

In order to simulate the micro-valve, the entire channel region is considered a fluid body, where Navier-Stokes equations for incompressible fluids are applicable. The bottom of the PDMS flap, which is the only solid entity, is fixed to the channel boundary (inner wall) and the rest of it is free to move under the influence of the force applied by the fluid velocity.

Figure 8 shows the surface, boundary and streamline plots of fluid velocity and the displacement of the PDMS flap due to fluid velocity. It can be noted from the figure that the fluid velocity is maximum in the region near the top of the obstacle. The bending of the PDMS obstacle, under the effect of fluid velocity load, is maximum at the top of the flap.

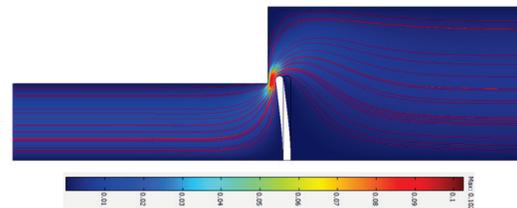


Figure 8. Surface and streamline plot of fluid velocity

As the fluid in the channel starts to flow under an applied pressure head, the PDMS flap tends to deflect depending on the direction of the flow. If the fluid is flowing from the narrow channel into the wider channel (from inlet to outlet), the flap would bend to the right and allow the fluid to pass. Conversely, if the fluid is flowing from the wider channel into the narrow channel (from outlet to inlet under the effect of back pressure), the flap would bend to the left and reduce the flow. If the deflection of the flap is large enough so that it touches the channel wall, it will completely block flow of fluid from the wider to the narrow channel. It was found

that the flow rate vs pressure variation is almost invariant with channel height ratio.

Figure 9 shows the variation of flow rate with pressure where the flap height is changed as a parameter. The graph depicts that higher flow rates can be obtained with smaller flap heights. Figure 9 reveals, under careful observation, that the nature of the flow rate vs. pressure plots almost resembles that of an electronic diode. Hence, it can be inferred that the micro-valve operates like a microfluidic diode, which allow fluid flow in one direction and blocks the flow in the opposite direction.

Depending on the type of PDMS used for the flap, its Young's modulus varies in the range 1-5 MPa. It is found out that the displacement of the flap decreases almost exponentially with the increase in the Young's modulus of the material.

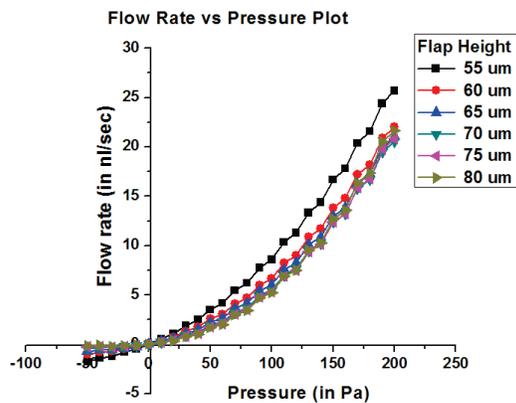


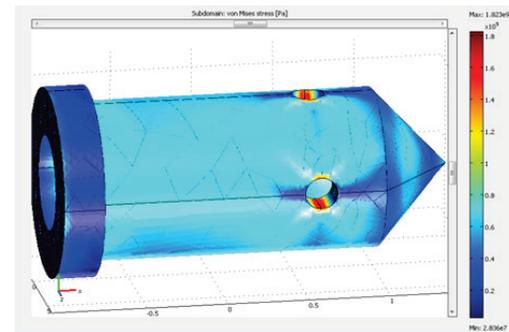
Figure 9. Plot of Flow Rate Vs. Pressure with Flap height as parameter

4.3 Simulation of the Micro-needle

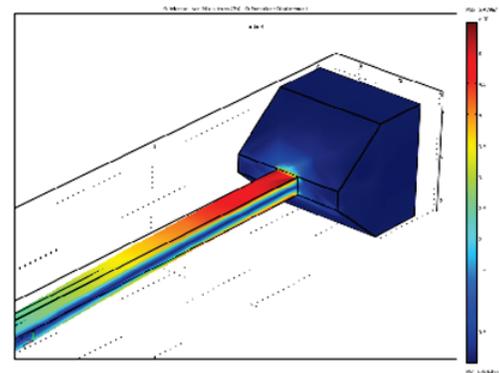
The maximum stress and deflection in the micro-needle structure are studied against different structural variables for both the out-of-plane and in-plane micro-needles. It is observed that under the influence of applied buckling force, the deformation of the structure is compressive in nature and the maximum stress is produced at the needle tips for the in-plane structure. But in case of the out-of-plane needle, the region of maximum stress is around the side-ports for the applied buckling force. However, on application of bending force in the lateral

direction, the deformation is deflective from the needle-axes and maximum stress is produced near the needle base for both the needles.

The deflection, maximum stress and fluid flow rates through the needle channel under a constant pressure head was plotted for different cross-sectional area of the needle bore. It can be easily noted from Figure 11 that the maximum stress and flow rate increases with increasing bore area.



(a)



(b)

Figure 10 (a) and (b). Region of maximum stress in the out-of-plane and in-plane needles for buckling and bending forces respectively.

In case of the out-of-plane micro-needle, a bore diameter of $40\mu\text{m}$ was chosen the and the flow rate is found to be almost $5\mu\text{l}/\text{second}$. A bore area of $864\mu\text{m}^2$ is chosen for the in-plane micro-needle, which allowed a flow rate of $0.064\mu\text{l}/\text{second}$. It was found that the deflection and maximum stress are within tolerable limits for both type of the micro-needles.

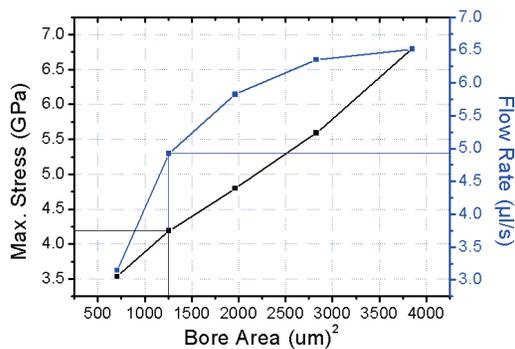


Figure 11. Variation of deflection, maximum stress and flow rate with needle bore area for out-of-plane needle

5. Conclusion

This paper addresses the issues related to the design and simulation of MEMS based piezoelectric micro-pump, micro-valve and silicon micro-needles. A piezoelectrically actuated 2-chamber micro-pump has been designed and simulated. The data obtained from the numerical analysis are used to calculate the volume flow rate of the device. It can be concluded that at a flow rate of 375 $\mu\text{l}/\text{min}$, the proposed micro-pump is capable of pumping microscopic fluid volumes. The flow rate can be adjusted by changing the operating frequency to suit specific application needs.

The PDMS flap based microvalve, a rather simple device, is capable of allowing fluid flow in one direction and blocking the flow in the opposite direction. This property can be exploited to prevent undesirable back flow of fluid in case of drug delivery and other such applications. The drugs can be delivered and the body fluids can be retrieved using the out-of-plane and in-plane micro-needles. Finally it was established from simulation results that the deflection, maximum stress and fluid flow rates were satisfactory for the chosen bore area of both the micro-needles.

The most important significance of the silicon substrate based micro-pump, micro-valve and the micro-needles is the possibility that these devices can be integrated to form a complete system capable of automatic sampling and drug delivery operations. Automated drug delivery systems like these, while being cheap, offer the

advantage of reliable operation without human intervention. In far-flung backward areas where the service of trained medical professionals are hardly available, these devices can play an important role in fulfilling the diagnostic and therapeutic needs of the local population.

6. References

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