Development of MEMS based Micropumps for Medical Application

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1 Introduction

Micropumps are essential components of the miniaturization of fluidic systems to enable liquid injection to systems and to control fluidic flow in a variety of applications such as integrated fluidic channel arrangements for chemical analysis systems or electronics cooling as well as for drug delivery systems. Micropumps offer important advantages because they are compact and small in size, they can operate using small sample volumes, and they provide rapid respond time. Miniaturized pump systems for chemical and biomedical applications have been widely studied. Various types of micropumps have been fabricated on different substrates such as peristaltic micropumps, metallic micropumps, plastic micropumps, as well as valveless piezoelectric micropumps. Among these types, valveless piezoelectric actuated micropumps have the advantage of moderate pressure and displacement at low power consumption, good reliability, and energy efficiency. They also respond rapidly and are widely used due to their ability to conduct particles without support from interior moving mechanical parts, thereby reducing the risk of clogging.

A number of different valveless micropumps employing nozzle-diffuser elements have been discussed in the literature. These include piezoelectrically actuated, electromagnetically actuated, and bubble micropumps. Use of nozzle-diffuser elements in magneto-hydrodynamic micropumps has also been reported. These pumps utilize the different pressure drop characteristics of flow through a nozzle and a diffuser to direct the flow in one preferential direction, and hence cause a net pumping action. Additional benefits of nozzle-diffuser elements include the ease of manufacture using conventional silicon micromachining techniques, and the much higher flow rates achievable with vibrating diaphragm pumps employing such valves. The higher flow rates, in spite of the poorer flow rectification properties of such valves, stem from the possibility of using valveless micropumps at much higher frequencies as compared to micropumps with passive check valves.

In this project, we are focusing on the application of valveless micropumps- nozzle/diffuser micropumps for medical applications like wound therapy. In literature, there are treatment for wounds- such as Vacuum Assisted Closure (VAC) or Negative Pressure Wound Therapy (NPWT). Our work provides a MEMS based micropump system which focuses on wounds not only outside the body, but also inside the body. A sensor is also incorporated to check the real time vitals of the patient.

2 Wound Therapy

Negative pressure wound therapy (NPWT) is a technique that enhances the healing process by applying negative pressure on the chronic or acute wounds. It has been diffusely adopted for treatment of trauma wound, chronic wound, or deep sternal wound infections due to its excellent healing result.
How does it work?

1. During the treatment, a device decreases air pressure on the wound. This can help the wound heal more quickly. The gases in the air around us put pressure over the area of the wound.
2. The therapy involves the controlled application of sub-atmospheric pressure to the local wound environment using a sealed wound dressing connected to a vacuum pump.
3. Vacuum therapy is a non-invasive massaging technique that helps to lift your skin via a mechanical device equipped with suction cups.

For the wound treatment using NPWT method, the wound environment is being controlled at the sub-atmospheric pressure. The sub-atmospheric pressure is usually lower than the atmospheric pressure, 760 mm Hg. This physics mechanics is able to induce mechanical stress to tissues and stimulate the division of cell (Mitosis). As such, the speed of growth of new blood vessels can be enhanced and wound will be drawn closed toward the center point. Thus, the speed of healing process by using negative pressure method will be faster and more efficient. Besides, NPWT method also enhances the healing process by increasing blood flow rate at wound area, maintaining the wound environment humidity, shielding the surrounding, reducing the risk of bacterial infection, decreasing interstitial oedema, contracting wound edges, and promoting the growth of granulation tissue. The whole description is shown in the figure below.

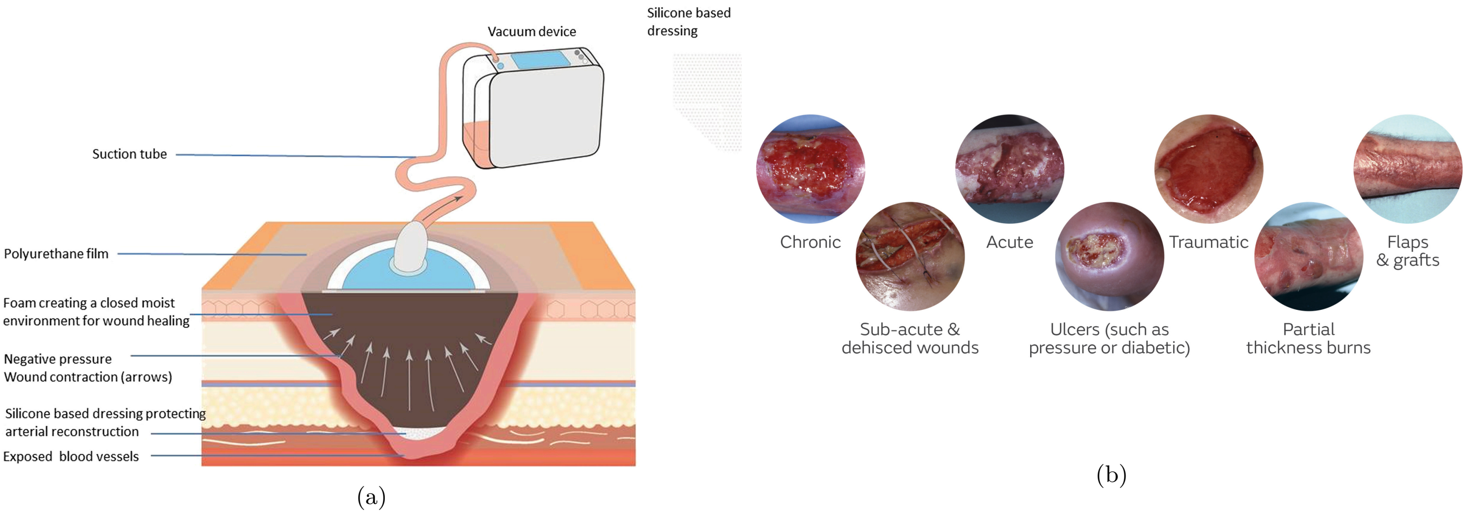


Figura 1: (a) Vaccum Assisted Closure(VAC) or Negative Pressure Wound Therapy(NPWT) system schematic (b) NPWT wound types

3 Target of the Project

We have seen in literature that even though there are several MEMS based devices in other medical aspects, but still there is a dearth of devices for wound therapy. As seen in classical NPWT, the main constituent of the VAC system is the pump which is creating the negative pressure over the wound. So, the first job is to design and simulate a micropump which can be applied not only to the wound outside the body, but also for cuts inside. A sensor will also be incorporated with the pump which will obtain real-time vitals of the person. After finalizing the design of both the micropump and the sensor, the whole device will be fabricated for further system integration.

4 Design of MEMS based wound therapy system

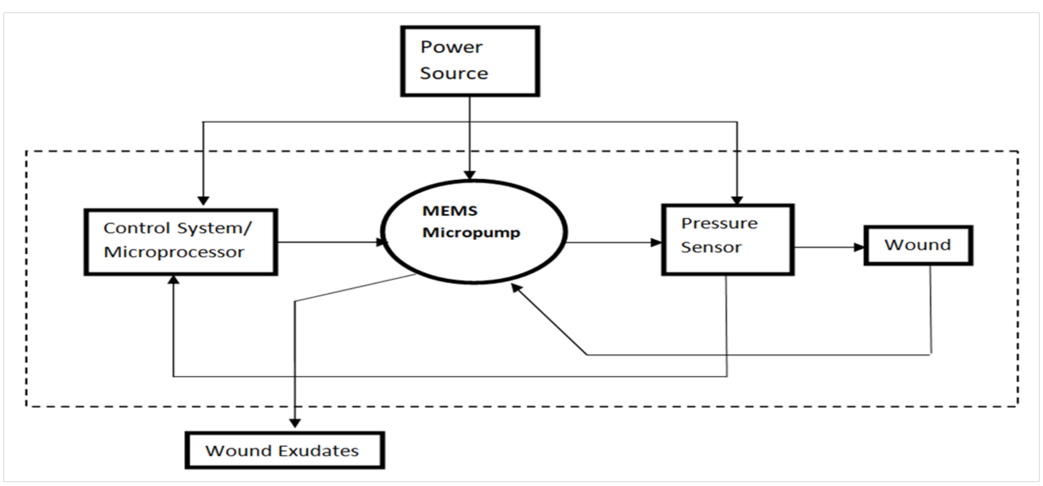


Figura 2: MEMS based wound therapy system

The overall wound therapy system design schematic is shown in fig 2. The system consists of a power source, the control system or the microcontroller, the pressure sensor and the MEMS micropump. The power source is applied to the micropump, microcontroller and the pressure sensor. To detect any unstable pressure generations at the wound, the pressure sensor is needed to monitor the change in pressure applied by the micropump. The sensor thus sends the signal to the microcontroller which controls the micropump. The micropump then would apply negetive pressure on the wound an remove the exudates out of the wound in a chamber.

4.1 Design of Micropump

4.1.1 Theoretical Analysis

- **Pressure Loss Co-efficient**

The pressure loss coefficient for flows through a gradually contracting nozzle, a gradually expanding diffuser, or a sudden expansion or contraction in an internal flow system is defined as the ratio of pressure drop across the device to the velocity head upstream of the device.

$$K = \frac{\Delta p}{\rho v^2/2} \tag{1}$$

For flow through a gradually expanding diffuser or a gradually contracting nozzle, the pressure loss coefficient can be calculated as follows. For flow in the diffuser direction, the incompressible steady-flow energy equation reduces to-

$$p_a + \frac{1}{2}\rho v_a^2 = p_b + \frac{1}{2}\rho v_a^2 + \Delta p_d \tag{2}$$

Hence, the pressure loss coefficient can be written as-

$$K_d = \frac{\Delta p_d}{\rho v_a^2/2} = \frac{p_a - p_b}{\rho v_a^2} + (1 - \frac{v_b^2}{v_a^2}) \tag{3}$$

Introducing the pressure recovery coefficient $C_p = \frac{p_b - p_a}{\rho v_a^2}$ and using the continuity equation $A_a v_a = A_b v_b$, K_d for spatial diffusers (e.g. conical and pyramidal) can be written as-

$$K_d = 1 - \frac{d_a^4}{d_b^4} - C_p \tag{4}$$

Hence for a given diffuser geometry, the pressure loss coefficient can be calculated from the pressure drop and the mean velocity at the neck. Similarly, for flow in the nozzle direction, the pressure loss coefficient is given by-

$$K_n = \frac{\Delta p_n}{\rho v_b^2/2} \tag{5}$$

- **Diffuser Efficiency**

The diffuser efficiency of a nozzle-diffuser element is defined as the ratio of the total pressure loss coefficient for flow in the nozzle direction to that for the flow in the diffuser direction.

$$\eta = \frac{K_{n,t}}{K_{d,t}} \tag{6}$$

Hence, $\eta > 1$ will cause a pumping action in the diffuser direction in a valveless micropump, while $\eta < 1$ will lead to pumping action in the nozzle direction. The case where $\eta = 1$ corresponds to equal pressure drops in both the nozzle and the diffuser directions, leading to no flow rectification. In Eq. (6), the total pressure loss coefficients for both the diffuser and nozzle directions can be divided into three parts: (i) losses due to sudden contraction at the entrance, (ii) losses due to gradual contraction or expansion through the length of the nozzle-diffuser and (iii) losses due to to sudden expansion at the exit. The total pressure drop in the diffuser direction can thus be written as-

$$\Delta p_{d,t} = \Delta p_{d,en} + \Delta p_d + \Delta p_{d,ex} \tag{7}$$

Therefore, the total pressure loss coefficient for the diffuser can be calculated as-

$$K_{d,t} = \frac{\Delta p_{d,t}}{\rho v_a^2/2} = \frac{\Delta p_{d,en}}{\rho v_a^2/2} + \frac{\Delta p_d}{\rho v_a^2/2} + \frac{\Delta p_{d,ex}}{\rho v_a^2/2} = K_{d,en} + K_d + K_{d,t}\frac{A_a^2}{A_b^2} \tag{8}$$

Similarly, the total pressure loss coefficient for the nozzle (with respect to pressure head at the neck) is-

$$K_{n,t} = (K_{n,en} + K_n)\frac{A_a^2}{A_b^2} + K_{n,ex} \tag{9}$$

Therefore, diffuser efficiency can be written as-

$$\eta = \frac{K_{n,t}}{K_{d,t}} = \frac{(K_{n,en} + K_n)\frac{A_a^2}{A_b^2} + K_{n,ex}}{K_{d,en} + K_d + K_{d,t}\frac{A_a^2}{A_b^2}} \tag{10}$$

- **Flow rectification efficiency**

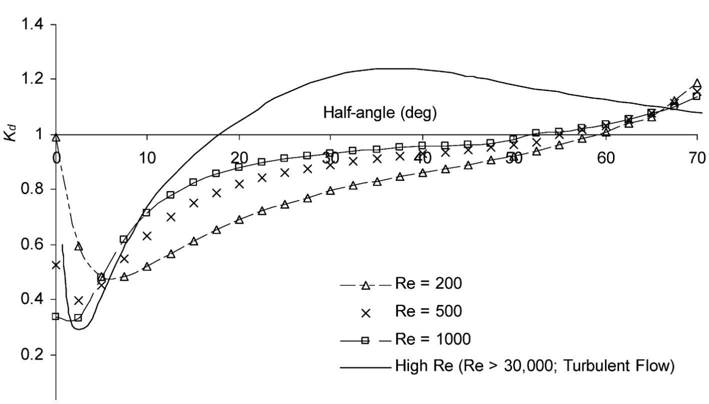
The flow rectification efficiency of a valveless micropump is the measure of the ability of the pump to direct the flow in one preferential direction. It can be expressed as-

$$\epsilon = \frac{Q_+ - Q-}{Q_+ + Q-} \tag{11}$$

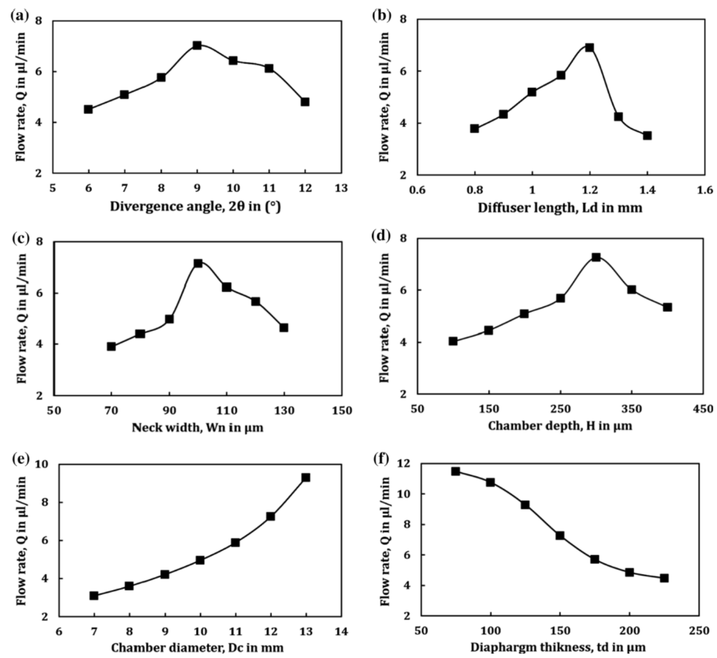
in which Q is flow rate and subscripts + and − refer to flow in the forward and the backward directions, respectively. A higher ϵ corresponds to better flow rectification. In particular, when there is no flow rectification, equal amounts of fluid move in both directions and $\epsilon = 0$, while for perfect rectification, flow is only in one direction and $\epsilon = 1$. The flow rectification efficiency of a valveless micropump is related to the diffuser efficiency of the nozzle-diffuser elements. As the diffuser efficiency departs from a value of 1, i.e. as the difference between $K_{n,t}$ and $K_{d,t}$ increases, ϵ for the micropump also increases.

4.1.2 Structure Schematic

The valveless micropump is a complex structure in a coupled fluidic system. The pump system consists of a piezoelectric actuator (piezo-disc), a silicon membrane, a pump chamber, and a microdiffuser/nozzle. Diffuser/nozzle elements, known as dynamic passive valves, are constructed on a planar surface. The diffuser/nozzle design determines the performance of the micropump. According to the theoretical analysis shown above, the structure of the micropumps is to be designed in such a way that the diffuser efficiency is as much high as possible.



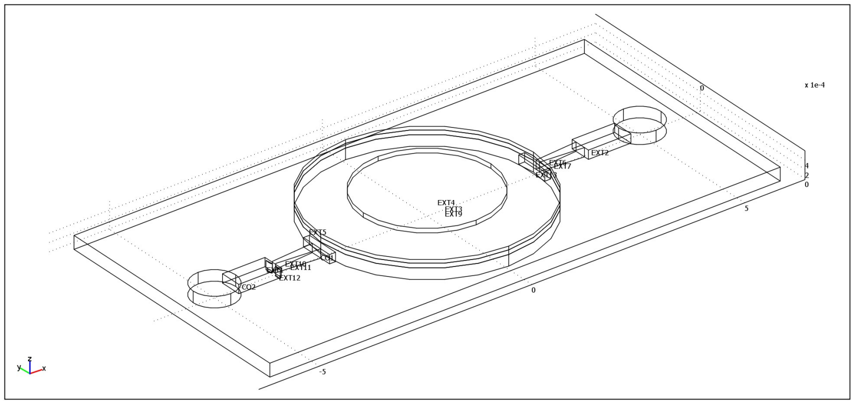
(a)



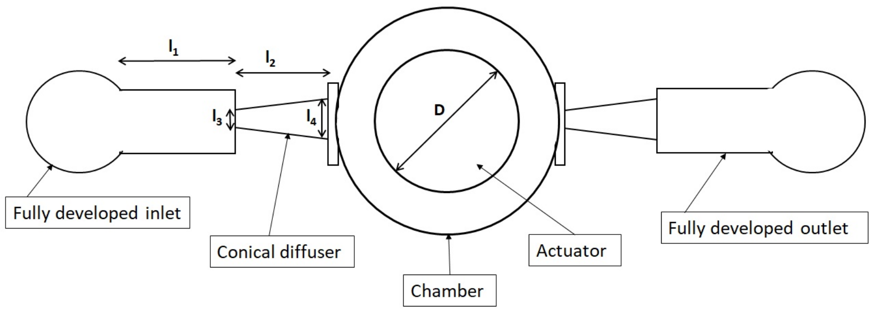
(b)

Figura 3: (a) Plot of variation of pressure loss coefficient in a conical diffuser with half-angle of the cone for Re = 200, 500, 1000 and >30,000 for fully developed inlet boundary layer(b) Parametric studies: flow rate as a function of (i) divergence angle, (ii) diffuser length, (iii) neck width, (iv) chamber depth, (v) chamber diameter and f diaphragm thickness, at actuation voltage of 50 V with actuation frequency of 100 Hz

According to previous literature, as shown in Fig 3(a), for the diffuser efficiency to be higher, the diffuser efficiency K_d should be much lesser. So for a conical diffuser, the pressure loss coefficient for a diffuser element was seen to low for low reynolds number at a half-angle of 4.5 ° - 5 ° . Therefore a diffuser angle of 9 ° was chosen for out design. In fig 3(b), the variation of flow rate with respect to different paramaters were shown and an optimal range of values were chosen for the design of the micropump. The micropump designed is $1.2 * 0.6mm^2$ as shown in Fig 4(a). The diameter of the chamber is $5mm$ with a depth of $250\mu m$. The thickness of the polyimide diaphragm is $80\mu m$ with a diamter of $5mm$. For the piezoelectric actuator, PZT-5H was used, with the material thickness of $100\mu m$ and diameter of $3mm$.



(a)



(b)

Figura 4: (a) Design of the proposed micropump. (b) Dimensions of the micropump.

The lengths of specific segments of the micropump is shown in fig 4(b). The designed conical diffuser has a length(l_2)of $1mm$. The length of the fully developed inlet channel had a length(l_1) of $1mm$. The neck width(l_3) was chosen $100\mu m$ and the outlet neck width was $260\mu m$.

4.1.3 Working Principle

The operating principle of a valveless micropump is illustrated in Fig. 5. The particular flow characteristics shown are for small nozzle-diffuser angles. In the expansion mode, as the volume of the pumping chamber increases, more fluid enters the pumping chamber from the element on the right which acts like a diffuser (and hence offers less flow resistance) than the element on the left, which acts like a nozzle. On the other hand, in the contraction mode, more fluid goes out of the element on the left which now acts as a diffuser, while the element on the right acts as a nozzle. Hence, net fluid transport is achieved in the pumping chamber from right to left.

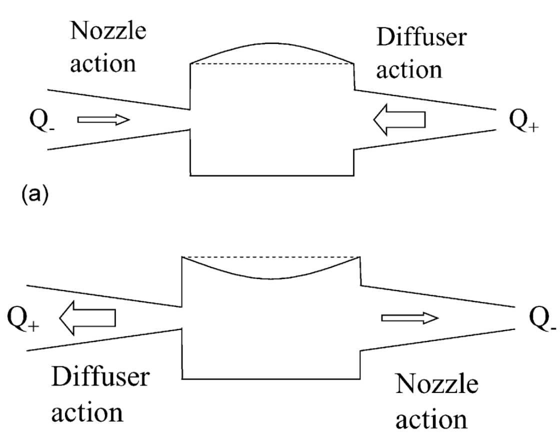
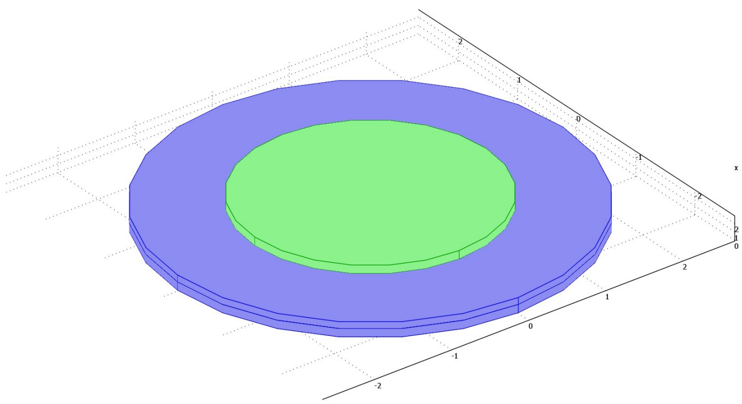


Figura 5: Working Principle of the micropump

5 COMSOL Simulation and Results

COMSOL Multiphysics is a cross-platform finite element analysis, solver and Multiphysics simulation software. It allows conventional physics based user interfaces and coupled systems of partial differential equations.
We designed the whole structure in MEMS module of COMSOL in Incompressible Navier Stokes. To divide the whole micropump two different simulations were done. First one was the piezoelectric actuator as a whole and then the fluid transport model in Microfluidics.

5.1 Piezoelectric Actuator



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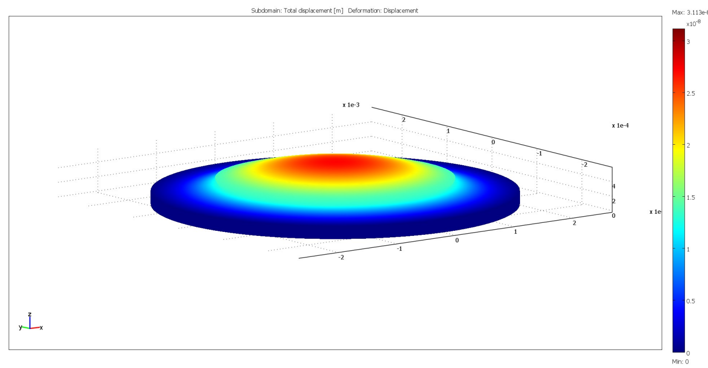

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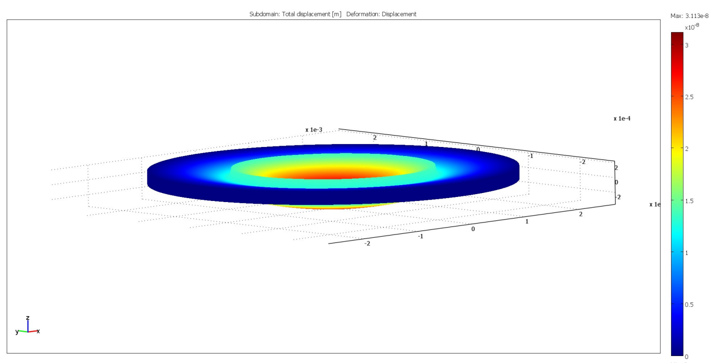

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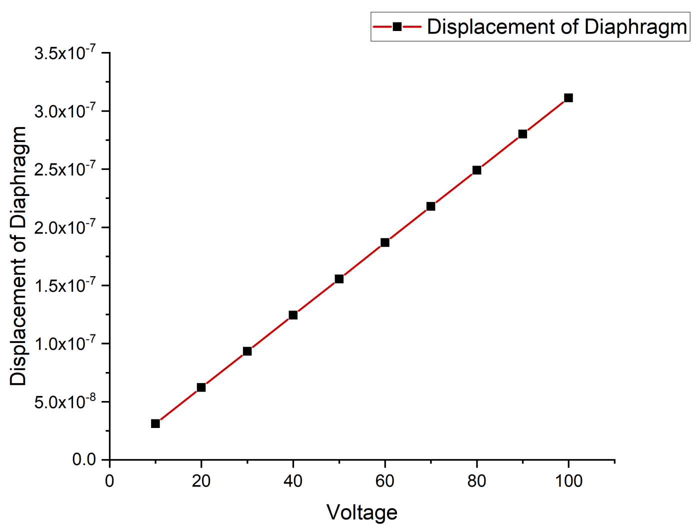
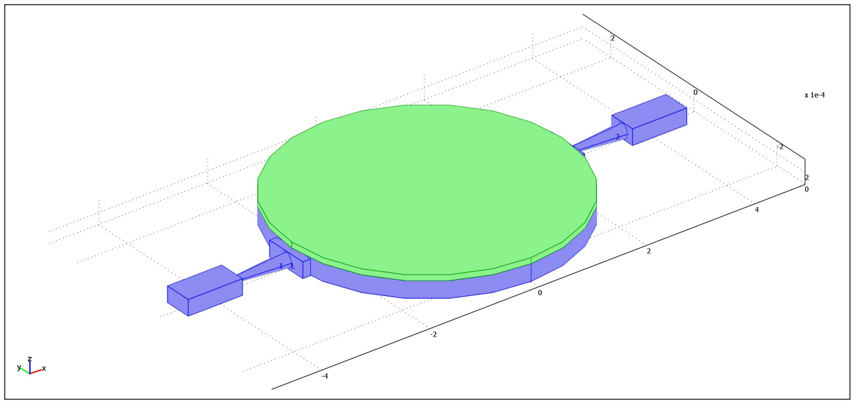


(d)

Figura 6: (a) Structure of the piezoelectric actuator(b) Actuator when a +10V is applied (c) Actuator when a -10V is applied (d) Displacement of the diaphragm vs. Voltage of the actuator

The piezoelectric actuator was simulated in COMSOL as shown in Fig 6. The diaphragm was selected to have a polyimide as the material, the base plate was selected to be brass and PZT-5H was selected to be the piezoelectric material for the actuator. It was seen when a positive potential was applied to the top surface of the actuator, the surface bends up and again when a negative voltage is applied, the surface bends down. This accounts for the pumping mechanism of the whole micropump. An analysis of displacement of the diaphragm vs. voltage has been done. It was seen in Fig 6(d), that the displacement increases linearly with the applied voltage.

5.2 Fluid Transport Geometry



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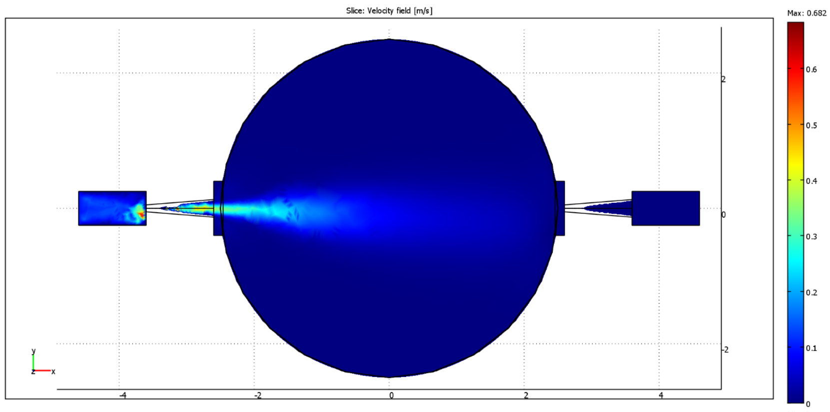

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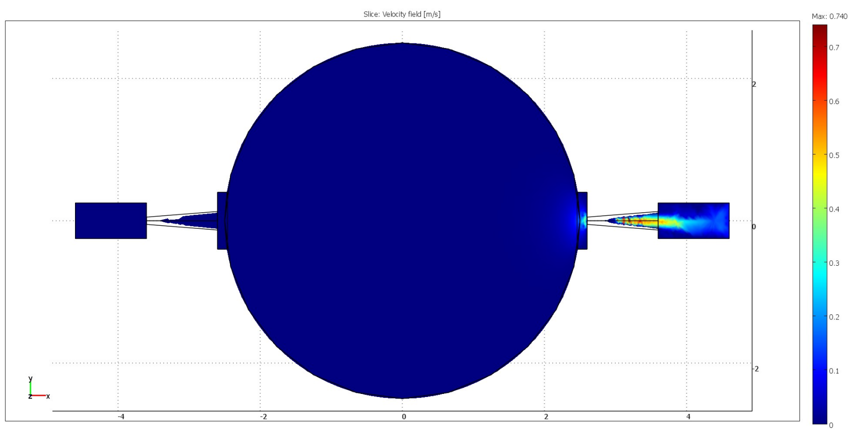


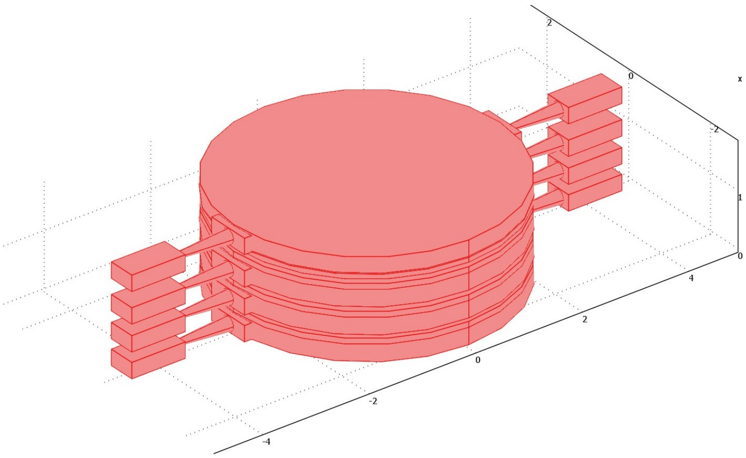
(c)

Figura 7: (a) Fluid Transport Geometry (b) Inflow characteristics with highest achievable velocity of 0.682m/s (c) Outflow characteristics with highest achievable velocity of 0.740m/s.

To analyze the fluid transport geometry, the structure was simulated in COMSOL. It was seen when a pressure was applied on the top plate of the pump, the maximum velocity for the inflow was observed around the neck of the diffuser with a value of 0.682m/s. For outflow, the maximum velocity of flow was observed along the length of the diffuser with a velocity of 0.740m/s.

6 Future Considerations

6.1 Design Improvement



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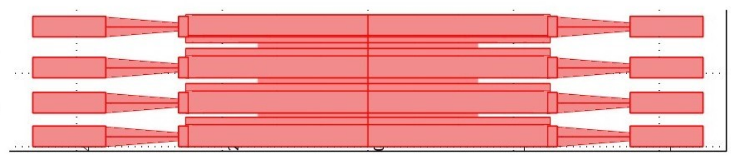

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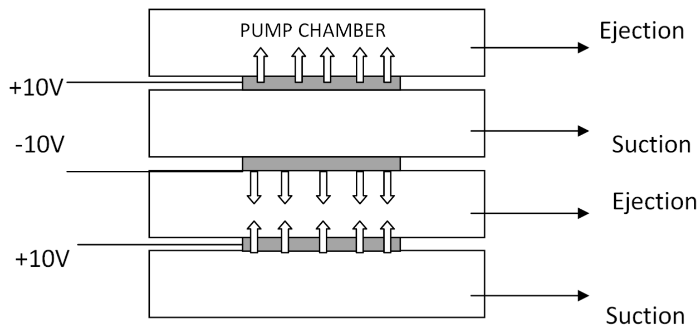

(c)

Figura 8: (a),(b) View of the future design of stacked micropumps, (c) Working Principle of the proposed micropump

We know it is imperative to have negative pressure over the wound for the therapy of NPWT. So, if only one micropump is used, application of constant pressure is not there. So, a design which will apply constant pressure on the wound is needed. One of such design improvement is shown in Fig 8. A system of four stacked pump is created with three piezoelectric actuator in the middle. The operation of the device is shown in Fig 8(c). If we apply a +10V to the top actuator, -10V to the middle actuator and +10V to the bottom actuator, we can see for the top substrate there will be ejection, for the second chamber there will be suction, for the third chamber there will be ejection and again for the bottom chamber there will be suction. So, we can achieve an alternate suction and ejection mode for the micropump and thus will be able to apply a constant pressure to the wound. If we think of flow rate, there will be increase in flow as there are no time interval between suction and ejection. Thus, building a much more efficient pump.
It was seen in previous literature that a rounded inlet for the neck of the diffuser element will increase the efficieny of the device. So, for other design improvement a rounded inlet will be made for the diffuser element.
A sensor will be incorporated with the micropump which will check the real-time vitals of the patient.

6.2 Fabrication

For the fabrication of the device, the process flow is as follows. Firstly, the a 300µm silicon wafer is patterned in the shape of the micropump with the corresponding chamber and two diffuser, inlet-outlet channels using photolithography. Then, it was etched out to form the required geometry. Isotropic etching is to be done for the conical diffusers and the chamber and the inlet boundary will be anisotropically etched. Then, the polyimide diaphragm was deposited on the wafer and patterned to give the specific shape. Then, the piezoelectric actuator with the brass base plate was kept on the diaphragm and joined with epoxy/glue resin.

6.3 Application Ranges

The whole system can have several application in medical domain. This device can be attached with the front of the imaging probe for endoscopy, and can attach to cuts inside the body after taking samples. For laparoscopy, cuts of 1-1.5cm is made. Our device's dimension is sufficient enough to mitigate the cut for such purpose.

7 Conclusion

In this project foundation we have discussed the design of the MEMS based micropump for medical applications such as wound therapy. We have designed and simulated the micropump using COMSOL Multiphysics
In the first section, we have designed a microfluidic micropump, which has been simulated, and experimentally analyzed.We have simulated the piezoelectric actuator and analyzed the change of displacement of the diaphragm with the applied voltage. We have also simulated the fluid flow characteristics inside the micropump which shows an inlet velocity of 0.682m/s and outlet velocity of 0.740m/s.
We have discussed the design improvements, steps of fabrication of the device and also the application ranges in medical domain.

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