Design and validation of a piezoelectric diaphragm micropump for drug delivery

Projeto e validação de uma microbomba piezoelétrica de diafragma para administração de medicamentos

Diseño y validación de una microbomba de diafragma de actuación piezoeléctrica para suministro de medicamentos

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ABSTRACT
There is a common eye ailment called Dry Eye Disease (DED) that causes symptoms of discomfort, visual disturbance, and tear film instability, which cause damage to the ocular surface. For treatment of this illness, the application of artificial tear fluid is usually prescribed. This work presents the design process of a piezoelectric micropump to become part of an automatic pumping system for delivering artificial tear fluid to people suffering from DED. Low polarizing voltage, portability, and achieving a variable output flow rate in the range of 1 µl/min, which is the necessary range needed by a person suffering from DED, are the focus of this design. Standard microfabrication techniques widely used in the CMOS industry and for MEMS development will be used for manufacturing the proposed micropump. For this reason, the materials and manufacturing processes involved must be CMOS-compliant. For the piezoelectric actuator, zinc oxide (ZnO) was selected due to its compatibility with CMOS technology and because it is readily available in thin film form through a sputtering process. Other materials involved in the device are silicon and aluminum. The design was validated through finite element analysis software, COMSOL Multiphysics®. The maximum flow rate obtained by a single micropump device was 2.51µl/min operating at 20V peak with a frequency of 900Hz. Also, the flow rate is controllable through variations in frequency, thus making it suitable for the application previously described.

Keywords: micropump, diaphragm, piezoelectric, drug delivery, zinc oxide.
RESUMO
Existe uma doença ocular comum chamada doença do olho seco (DED), que causa sintomas de desconforto, distúrbios visuais e instabilidade do filme lacrimal, que causam danos à superfície ocular. Para o tratamento desta doença, geralmente é prescrita a aplicação de fluido lacrimal artificial. Este trabalho apresenta o processo de projeto de uma microbomba piezoelétrica para fazer parte de um sistema de bombeamento automático para fornecimento de fluido lacrimal artificial a pessoas que sofrem de DED. Baixa tensão de polarização, portabilidade e obtenção de uma vazão de saída variável na faixa de 1 µl/min, que é a faixa necessária para uma pessoa que sofre de DED, são o foco deste projeto. Técnicas padrão de microfabricação amplamente utilizadas na indústria CMOS e para desenvolvimento de MEMS serão utilizadas para a fabricação do projeto proposto. Por esta razão, os materiais e processos de fabricação envolvidos devem ser compatíveis com CMOS. Para o atuador piezoelétrico, o óxido de zinco (ZnO) foi selecionado devido à sua compatibilidade com a tecnologia CMOS e porque está prontamente disponível na forma de filme fino através de um processo de pulverização catódica. Outros materiais envolvidos no dispositivo são silício e alumínio. O projeto foi validado através de simulações de elementos finitos utilizando o software COMSOL Multiphysics®. A vazão máxima obtida por um único dispositivo de microbomba foi 2.51µl/min, operando em pico 20V com frequência de 900Hz. Além disso, a vazão é controlável através de variações de frequência, tornando-a adequada para a aplicação descrita anteriormente.

Palavras-chave: microbomba, diafragma, piezoelétrica, drug delivery, óxido de zinco.

RESUMEN
Existe una dolencia ocular común llamada enfermedad del ojo seco (EOS) que causa síntomas de malestar, alteración visual e inestabilidad de la película lagrimal, que provocan daños en la superficie ocular. Para el tratamiento de esta enfermedad se suele prescribir la aplicación de lágrimas artificiales. Este trabajo presenta el proceso de diseño de una microbomba piezoeléctrica para que forme parte de un sistema de bombeo automático para administrar líquido lagrimal artificial a personas que padecen EOS. El enfoque de este diseño es el bajo voltaje de polarización, la portabilidad y el logro de un caudal de salida variable en el rango de 1 µl/min, que es el rango necesario para una persona que sufre de EOS. Para fabricar el diseño propuesto se utilizarán técnicas de microfabricación estándar ampliamente utilizadas en la industria CMOS y para el desarrollo de MEMS. Por este motivo, los materiales y procesos de fabricación implicados deben ser compatibles con CMOS. Para el actuator piezoeléctrico, se seleccionó óxido de zinc (ZnO) debido a su compatibilidad con la tecnología CMOS y porque está disponible en forma de película delgada mediante un proceso de pulverización catódica. Otros materiales involucrados en el dispositivo son el silicio y el aluminio. El diseño fue validado mediante simulaciones de elementos finitos utilizando el software COMSOL Multiphysics®. El caudal máximo obtenido por un solo dispositivo de microbomba fue 2.51µl/min, operando en el pico 20V con una frecuencia de 900Hz. Además, el caudal es controlable mediante variaciones de frecuencia, lo que lo hace adecuado para la aplicación descrita anteriormente.

Palabras clave: microbomba, diafragma, piezoeléctrica, administración de medicamentos, óxido de zinc.
1 INTRODUCTION

The development of new pharmaceutical products that improve patients’ health was enabled by emerging drug delivery technologies. This was done by enhancing the delivery of a therapeutic drug to its target site, minimizing off-target accumulation, and facilitating patient compliance. For all drugs, the goal of delivery is to maximize therapeutic efficacy by transporting and releasing the drug to the target site in the body at the correct dose (Vargason et al., 2021). These new pharmaceutical products have a wide range of physiochemical and pharmacokinetic properties that conventional (oral or intravenous) administration might not be able to deliver effectively, thus requiring targeted drug delivery with controlled release and making use of microscale delivery devices to enable site-specific targeted drug delivery (Lee et al., 2018). The use of microelectromechanical systems (MEMS) in drug delivery systems (DDS) has allowed the development of new delivery systems capable of overcoming the challenges that arise in a vast variety of applications. The main advantages of MEMS include miniaturization, multiple function integration, and electromechanical control; this allows for localized drug delivery to challenging areas in the human body, as in the case of the eye (Joshita et al., 2017).

Micropumps are microfluidic devices capable of delivering a controlled flow (usually in µl) through the microchannels connected to them. They find their application in DDS by transporting the proper dose of the drug from the reservoir to the target site in the body, thus improving the effect of treatment on patients’ health (Shawgo et al., 2002; Mohit et al., 2019). Early designs of micropumps were intended to deliver insulin to diabetes patients (vanLintel et al., 1988; Sefton et al., 1981), starting the use of these devices in DDS. Micropumps also have many applications, such as micro-total analysis systems (µTAS), space exploration with micro-spacecraft, and heat dissipation in space-constrained electronics (Mohit, 2019).

According to the International Dry Eye Workshop (DEWS, 2007), dry eye is a multifactorial disease of the tears and ocular surface that results in symptoms of discomfort, visual disturbance, and tear film instability, with potential damage to the ocular surface. This is a common disease that can be chronic or episodic, and risk factors for suffering it include smoking, alcohol consumption, age, and ethnicity (Graue-Hernandez et al., 2018, Aires Sampaio et al., 2021). The best treatment for DED is the direct application of artificial tear supplements to improve symptoms, improve tear-film
stability, and reverse ocular surface damage (O’Brien et al., 2004). However, it has been found that this treatment only temporarily relieves the symptoms of dry eye rather than healing the ocular surface or treating the underlying cause of the disease, causing the need for continual application of the treatment (Gayton, 2009, Mattos Fernandes et al., 2023).

For treating this illness, the most common DDS used is the manual application of artificial tears to the ocular surface to keep it protected. Although a good treatment method, some drawbacks are: the unreliability of dropper tips for delivering the correct dose (German et al., 1999); and the difficulty found in some populations suffering from this ailment for self-administration of the treatment, as in children and elderly people. As a result, new automated devices for administering the treatment were devised to solve these issues (Meni-Babakidi et al., 2020; Murube et al., 2003; Pillay, 2016).

In this paper, a piezoelectric valveless diaphragm micropump to aid in the treatment of DED is presented, focusing on low voltage polarization, small size, and flow controllability. This design is meant to be a part of an automated device for administering artificial tear fluid, so it was designed to work with that kind of fluid. The device is meant to be manufactured using standard microfabrication techniques used in MEMS, thus allowing further integration of integrated control circuitry and solid-state sensors within the device. In the following sections, an outline of the design workflow used in the creation of the design and the validation of the design by finite element analysis software will be presented.

2 WORKING PRINCIPLES INVOLVED IN THE MICROPUMP OPERATION

The operation of a common diaphragm micropump is displayed in Figure 1. This process can be divided into two phases: the suction phase and the pumping phase. During the suction phase, the diaphragm is forced to bend upwards, creating an increase in the internal volume of the pumping chamber that is filled with the working fluid coming in from the inlet. After that, in the pumping phase, the diaphragm bends downward, causing a decrease in the volume of the pumping chamber and pushing the working fluid to the outlet of the device. The cyclical repetition of these two phases will create a controlled average flow rate from inlet to outlet (Joshita et al., 2019).
Figure 1. Working principle of the micropump. The blue arrow shows the bending direction of the diaphragm, while the red arrow indicates the fluid flow direction.

Source: Prepared by the authors themselves.

Most diaphragm micropumps make use of a transducer to convert the supplied energy (usually electrical) into a mechanical force for bending the diaphragm. The actuators can be electrostatic, magnetostatic, thermo-pneumatic, piezoelectric, etc. In the last case (Figure 2), the piezoelectric material is attached to the clamped diaphragm, and then it is polarized with an alternating voltage, so it starts suffering from cyclical internal mechanical tensions (the indirect piezoelectric effect). These tensions cause the diaphragm to start oscillating, which causes the necessary changes in volume inside the pumping chamber required for the liquid to start flowing (Mohit et al., 2019).

Figure 2. Operation of a piezoelectric diaphragm.

Source: Prepared by the authors themselves.

Another important aspect to deal with is flow rectification, which involves making the flow follow a defined preferred direction. In a micropump, the flow must be from inlet to outlet. To achieve this, a set of valves is attached to the input and output of the pumping chamber to block the flow of fluid in the opposite direction. The most common type of microvalves are nozzle/diffuser elements; the micropumps that use this configuration are also called "valveless." These kinds of valves are microfluidic channels that gradually shrink (or enlarge) along their length. The operation of these valves is as follows: during
the suction phase, the sudden increase in volume of the pumping chamber causes fluid to go inside; the right valve acts as a diffuser, letting the fluid go from the input to inside the pumping chamber; and the left valve, which is acting as a nozzle, opposes the flow to the pumping chamber. During the pumping phase, the diaphragm exerts its pressure on the fluid inside the pumping chamber, causing the roles of the valves to exchange. This is pictured in Figure 3.

Figure 3. Nozzle/diffuser element (left) and operation (right).

Source: Prepared by the authors themselves.

3 DESIGN WORKFLOW

Figure 4 shows the general workflow for designing the piezoelectric micropump, following the systems engineering method (Kossiakoff et al., 2011). The process starts with the requirement analysis, in which the available input conditions and desired outputs are defined according to the necessities of the problem to solve. The process follows the functional definition; in this step, the requirements are translated into functions that the system must accomplish, the system elements are partitioned, and the interaction between these elements is defined.

Figure 4. Workflow design of the micropump.

Source: Prepared by the authors themselves.

Next, in the physical definition step, the most adequate design approach is selected by the trade-off of prioritized criteria for obtaining the best compromise among...
performance, risk, cost, etc. Then, the design is elaborated to the necessary level of detail. Finally, the design is validated through mathematical models of the system environment that reflect all the requirements and constraints, then simulating system solutions and iterating as necessary to obtain system solutions similar to the desired ones. The end result of this process is a design that complies with the established requirements for solving the problem at hand.

3.1 REQUIREMENT ANALYSIS

For the design of a piezoelectric micropump capable of supplying artificial tear fluid for the treatment of symptoms of DED, the design must be capable of transporting an adequate dose from the reservoir to the person’s eyes.

In a healthy individual, the rate at which the eye produces tears is 1.10 µl/min. In the case of people suffering from DED, this flow rate could be reduced by half or even further (Cerretani, 2014). Thus, the design must be able to pump artificial tear fluid to the person’s eyes at a flow rate in that range for supplying the treatment. This flow rate must also be controllable to accommodate the required doses for different patients.

Another important aspect of the device is its dimensions. Since it is meant to be part of an automatic pumping system for delivering artificial tear fluid, the overall device dimensions should be as small as possible. The main feature of modern DDS is that they can be placed as close as possible to the area to treat.

Figure 5. Schematic of the human eye.

Source: Adapted from the Mayo Clinic web site, Dry Eyes, URL: https://www.mayoclinic.org/diseases-conditions/dry-eyes/symptoms-causes/syc-20371863
Figure 5 shows a picture of the human eye and the lacrimal system. Given that the eye’s diameter is 24mm (Gross, 2008), the whole device must be within that range or smaller to place it as close as possible to the eye for supplying the required artificial fluid.

Finally, since the device is meant to be portable, its power supply will also be small in size and capacity. These constraints require the device to operate with the least amount of energy possible in order to work for a considerable amount of time. In brief, the proposed device should focus on: achieving a controllable flow rate within the 1µl/min range, compact size, and low energy consumption.

3.2 FUNCTIONAL DEFINITION

To address these needs, the design will make use of standard microfabrication techniques used in MEMS for its manufacture. These widely used techniques in the manufacturing of integrated circuits and MEMS devices will ensure the small size of the device, low power consumption, and reliability of the micropump. The general construction of a piezoelectrically actuated diaphragm is shown in Figure 6. For this design, the diaphragm will be made up of crystalline silicon, which is the most commonly used material in the semiconductor industry. For the piezoelectric actuator, the chosen material will be zinc oxide since it’s a CMOS-compliant material available in thin films through sputtering deposition. The piezoelectrically actuated diaphragm is constructed as follows: the bottom layer is that of the silicon diaphragm, followed by the tri-layer construction of the piezoelectric actuator, a thin film of aluminum as the bottom electrode, followed by a thin film of zinc oxide as the piezoelectric material, and a thin film of aluminum as the top electrode.

![Figure 6. Structure of a piezoelectrically actuated diaphragm.](image)

Source: Prepared by the authors themselves.
3.3 PHYSICAL DEFINITION

For the manufacture of the micropump, the overall design must be planned in accordance with the microfabrication techniques used in the industry, such as photolithography, thin film deposition, volume micromachining, wet etching, etc.

The main structure of the micropump will be made of three separated substrates (Figure 7). The first one will be a crystalline silicon substrate with the piezoelectrically actuated diaphragm placed on its top face. The next substrate will also be a crystalline silicon wafer with the microfluidic microchannels etched by bulk micromachining. Finally, a glass substrate with openings for inlet and outlet ports will be used as the mechanical support of the device. All three of these substrates will be bonded to form the complete device.

In this design, the diaphragm will be created by the micromachining of cavities in the silicon substrate, and the piezoelectric actuator will be created by the deposition of an aluminum film, followed by a zinc oxide film, and finally another aluminum film. The two aluminum films work as electrodes for polarizing the piezoelectric material. There is a wide variety of piezoelectric materials, but zinc oxide was selected because of its high compatibility with CMOS technology and low toxicity compared to lead-containing piezoelectric materials such as PZT.

In this design, the diaphragm will be created by the micromachining of cavities in the silicon substrate, and the piezoelectric actuator will be created by the deposition of an aluminum film, followed by a zinc oxide film, and finally another aluminum film. There is a wide variety of piezoelectric materials, but zinc oxide was selected because of its high compatibility with CMOS technology, its low toxicity compared to lead-containing piezoelectric materials such as PZT, and the fact that it can be easily obtained through the sputtering process. Finally, the working fluid taken into consideration for the
The development of this work was a solution of synthetic tears called Thealoz®. Figure 8 shows the schematics of the micropump design; Table 1 lists its measurements; and Table 2 lists the physical properties of the materials involved.

Table 1. Measurements of the proposed micropump.

<table>
<thead>
<tr>
<th>PARAMETER</th>
<th>VALUE</th>
<th>UNIT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Micropump length ((L_d))</td>
<td>8</td>
<td>mm</td>
</tr>
<tr>
<td>Micropump width ((W_d))</td>
<td>4</td>
<td>mm</td>
</tr>
<tr>
<td>Diaphragm diameter ((d_p))</td>
<td>3</td>
<td>mm</td>
</tr>
<tr>
<td>Piezoelectric diameter ((d_{pz}))</td>
<td>2.4</td>
<td>mm</td>
</tr>
<tr>
<td>Piezoelectric length ((l_{pz}))</td>
<td>1.8</td>
<td>mm</td>
</tr>
<tr>
<td>Diaphragm thickness ((t_p))</td>
<td>50</td>
<td>µm</td>
</tr>
<tr>
<td>Electrode thickness ((t_e))</td>
<td>300</td>
<td>nm</td>
</tr>
<tr>
<td>Piezoelectric thickness ((t_{pz}))</td>
<td>100</td>
<td>nm</td>
</tr>
<tr>
<td>Nozzle/Diffuser length ((l_{nd}))</td>
<td>1.2</td>
<td>mm</td>
</tr>
<tr>
<td>Nozzle/Diffuser small opening ((a_{nd}))</td>
<td>80</td>
<td>µm</td>
</tr>
<tr>
<td>Nozzle/Diffuser wide opening ((A_{nd}))</td>
<td>460.12</td>
<td>µm</td>
</tr>
<tr>
<td>Thickness of the micropump channels ((t_{ch}))</td>
<td>300</td>
<td>µm</td>
</tr>
</tbody>
</table>

Source: Prepared by the authors themselves.

Table 2. Measurements of the proposed micropump.

<table>
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<th>MATERIAL</th>
<th>PARAMETER</th>
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<tr>
<td>Thealoz®</td>
<td>Density</td>
<td>1000</td>
<td>kg/m³</td>
</tr>
<tr>
<td></td>
<td>Dynamic Viscosity</td>
<td>1.047</td>
<td>mPa·s</td>
</tr>
<tr>
<td>Crystalline Silicon</td>
<td>Density</td>
<td>2330</td>
<td>kg/m³</td>
</tr>
<tr>
<td></td>
<td>Elasticity matrix</td>
<td></td>
<td>GPA</td>
</tr>
<tr>
<td></td>
<td>Relative permittivity</td>
<td>11.7</td>
<td>1</td>
</tr>
<tr>
<td>Aluminum</td>
<td>Density</td>
<td>2700</td>
<td>kg/m³</td>
</tr>
<tr>
<td></td>
<td>Young Modulus</td>
<td>70</td>
<td>GPA</td>
</tr>
<tr>
<td></td>
<td>Poisson’s Ratio</td>
<td>0.35</td>
<td></td>
</tr>
</tbody>
</table>

Source: Prepared by the authors themselves.
### 3.4 DESIGN VALIDATION

In order to validate the design, the finite element analysis software COMSOL Multiphysics® was used. For the simulations of the device, three physics modules were used: solid mechanics, electrostatics, and fluid flow. Also, two multiphysics couplings were used: the piezoelectric effect and the fluid-structure interaction. The former couples the solid mechanics and electrostatics modules, while the latter couples the solid mechanics with the fluid flow module. In this way, it is possible to simulate the mechanical effect generated by the piezoelectric diaphragm and how it transfers to the working fluid in order to pump it through the microfluidic channels.

![3D geometry](source)

Source: Prepared by the authors themselves.

The 3D geometry used for the simulation of the device is shown in Figure 9. In the 3D geometry, only the fluid volume (blue color on the figure) and the actuator volume

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Zinc Oxide

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Relative permittivity</td>
<td>8.54</td>
</tr>
<tr>
<td>Density</td>
<td>5680 kg/m³</td>
</tr>
</tbody>
</table>

**Elasticity matrix**

\[
\begin{bmatrix}
210 & 121 & 153 & 0 & 0 & 0 \\
121 & 297 & 153 & 0 & 0 & 0 \\
153 & 153 & 211 & 0 & 0 & 0 \\
0 & 0 & 0 & 42 & 0 & 0 \\
0 & 0 & 0 & 0 & 42 & 0 \\
0 & 0 & 0 & 0 & 0 & 44 \\
\end{bmatrix} \quad \text{GPa}
\]

**Coupling Matrix**

\[
\begin{bmatrix}
0 & 0 & 0 & 0 & -11.3 & 0 \\
0 & 0 & 0 & -11.3 & 0 & 0 \\
-5.43 & -5.43 & 11.7 & 0 & 0 & 0 \\
\end{bmatrix} \quad \text{pC/N}
\]

Source: Prepared by the authors themselves.
that will be in contact with the fluid (gray and purple regions) were drawn and set for simulation. The boundary conditions applied were set according to the physics module. For the solid mechanics module, the lateral area of the diaphragm border was set as a fixed boundary (dashed green line around the diaphragm), allowing for the bending of the diaphragm along its vertical axis. For the electrostatics module, the bottom electrode was set as ground terminal (GND), while the top electrode was polarized with a voltage of $V_{in} = 20 \sin(2\pi ft) \, V$. Finally, for the fluid mechanics module, the two openings of the design were set as fluid inlet and outlet at 0Pa, as indicated by Figure 9. Finally, the physics couplings were defined, and the simulations were computed for 30 operation cycles, varying the frequency from 100Hz to 1300Hz.

4 RESULTS AND DISCUSSIONS

After carrying out the simulations, the average flow rate of the micropump was obtained for each frequency value. The obtained plot for the flow rate of each frequency value resembles that of Figure 10. The plot shows an oscillating tendency that correlates to the sinusoidal applied voltage. This is just the instantaneous flow rate produced by the pump.

For obtaining the average flow rate, first the difference between the flow rate at the outlet and inlet was computed, then an average of this difference was calculated. The average flow rate of the micropump for every frequency value evaluated is shown in
Figure 11. As can be seen from the picture, the 1µl/min milestone was reached at a frequency of 900Hz while being operated by a voltage of $V_{in} = 20 \sin(2\pi ft) V$.

Figure 11. Average flow rate of the fluid at different frequencies.

![Graph of average flow rate vs frequency](source)

In order to get a wider insight into the results of the simulation, we take the period of time of the applied voltage (Figure 12) between the points 0.0312s and 0.0322s. The maximum positive voltage (or peak) is located at time 0.31389 s, while the lowest negative voltage applied (or valley) is situated at point 0.31944.

Figure 12. One period of the operation voltage applied to the actuator.

![Graph of operation voltage](source)

As can be seen in Figure 13, the maximum upward bending of the diaphragm occurs just when the maximum possible voltage (20 V) is applied. On the other hand, the lowest downward bendig was achieved during the valley of the negative semi-cycle of the polarizing voltage. This is the expected behavior of the diaphragm, just as it was described before.
Figure 13. 3D plot of the displacement suffered by the diaphragm at the peak and valley of the sinusoidal applied voltage.

Source: Prepared by the authors themselves.

The displacement plot of the diaphragm resembles that of Figure 14. It can be noticed that the displacement is a periodic function, just as the input voltage is, and that the magnitudes of the maximum and minimum values reached are practically the same. Then, for every frequency value, the maximum displacement was obtained; the results are shown in Figure 15. To understand these results, the first-mode resonance frequency was computed and calculated at 85.5kHz.

Figure 14. Displacement suffered by the diaphragm at 900 Hz.

Source: Prepared by the authors themselves.
Figure 15. Maximum displacement of the diaphragm at different frequencies.

Source: Prepared by the authors themselves.

Figure 16 shows the flow profile across the microchannels; every arrow is a defined point taken for this representation, and the length of the arrow represents its velocity magnitude. Taking into consideration the same operating period as before, it can be seen that during the peak of the input voltage, fluid is coming in from the inlet while little water is coming in from the outlet. During the valley of the operating voltage, the operation changes; now more fluid is getting out of the outlet than from the inlet.

Figure 16. 3D plot of the flow profile at the peak and valley of the sinusoidal applied voltage.

\[ \text{Source: Prepared by the authors themselves.} \]
The results above showed that the piezoelectric micropump can achieve flow rates within \(1\mu l/min\), which was a requirement for this work. Also, the flow rate produced by the device is controllable through variations in frequency. As expected, the maximum displacement achieved by the micropump increases with an increase in the frequency. This is because the more the diaphragm operates at a frequency near the resonant frequency of the piezoelectric diaphragm, the maximum displacement increases. This means that the micropump could produce higher flow rates by increasing the operation frequency, although it will be limited by the response time limit of the fluid or the piezoelectric diaphragm.

Finally, the flow profiles show the expected fluid flow inside the microchannels. As can be seen by the flow rate plot from Figure 10, the micropump is capable of producing an instant flow greater than \(10\mu l/min\), but it gets reduced when obtaining the average net flow. This is because on every bend of the diaphragm, fluid is suctioned or pumped through both openings of the pump chamber, but in different quantities, which, if taken for a long enough period of operation, would average to a constant net flow rate from input to output. The maximum flow rate of the micropump can be increased by redesigning the valves to achieve higher rectification or by changing them for check valves, which are more efficient. In this work, nozzle/diffuser valves were favored because of their ease of manufacturing with microfabrication techniques.

5 CONCLUSION

In this work, a piezoelectric micropump to become part of an automatic pumping system for delivering artificial tear fluid to people suffering from DED was designed with the following requirements: low operating voltage, controllable flow rate, and compact size. The micropump has a controllable flow rate that goes from \(0.02\mu l/min\) at 100\(Hz\) to \(2.51\mu l/min\) at 1300\(Hz\). The \(1\mu l/min\) milestone was achieved at 900\(Hz\) and several lower values within this range can be obtained through variations in the frequency, thus accommodating for the requirements of different patients. Finally, its polarizing voltage is a sinusoidal wave with a peak of 20 \(volts\).
REFERENCES


