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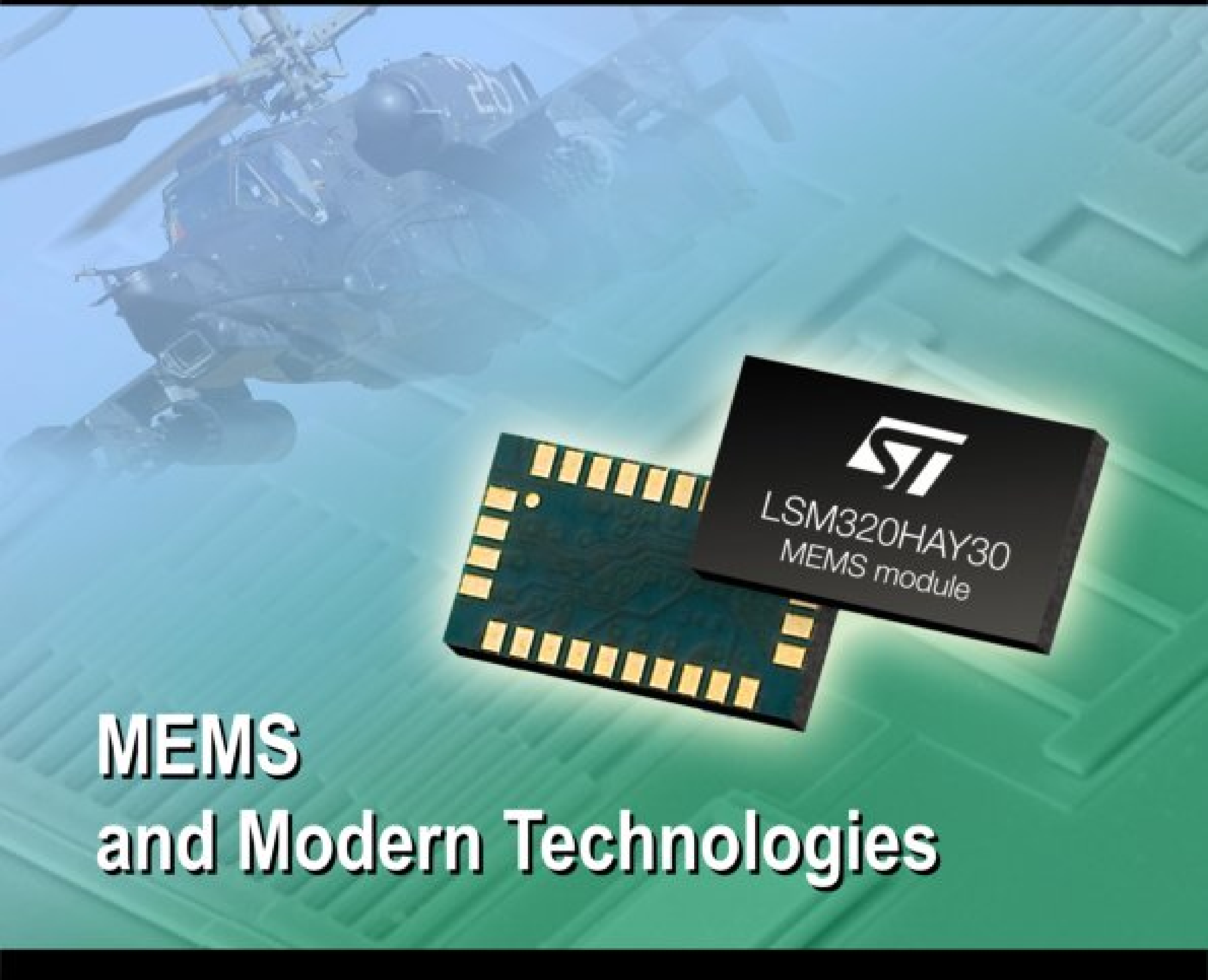

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Simulation and Design Optimization of Piezoelectricaly Actuated Valveless Blood Pump for Hemofiltration System

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Abstract: Kidney failure is expanding almost exponentially in the world at present. So, there is an urgent need to develop highly efficient and low cost system capable of blood filtration for the patients who need to remove the waste products from their blood to save their lives. This paper presents a novel design of MEMS based hemofiltration (HF) system. The proposed HF system consists of four modules that are electronic module, blood transport module, blood filtration module and drug delivery module. Coupled multifield analysis of MEMS based piezoelectrically actuated polymeric valveless micropump has been performed for HF system. Micropump comprises of thee serial actuators, nozzle/diffuser elements and three chambers. Effects of various parameters such as actuator thickness, frequency, voltage, nozzle/diffuser angle, power consumption on flow rate have been investigated during the multifield and theoretical analysis. The maximum deflection of 6.808 μm has been observed at applied 105 V. Maximum flow rate of 519 μL/min has been obtained at 200 Hz with applied 105 V by keeping thickness constant 0.2 mm with power consumption 45 mW.

Keywords: Hemofiltration system, MEMS, Piezoelectric micropump, Coupledmultifield.

1. Introduction

Microfluidics is relatively a new and advanced field of science and technology. In last few years, micro and nano electromechanical system (MEMS and NEMS) based microfluidic systems have revealed their potential for biomedical applications [1]. In literature, a range of microfluidic systems have been presented for many biomedical applications like drug delivery, genetic analysis, blood transport, blood filtration, biochemical analysis, cell separation, sample preparation, chemical synthesis, drug screening, electro chromatography, etc. Most important components of the MEMS based microfluidic systems are micropumps, microchannels, microneedles, check valves, micro-chambers, micro-mixers, micro-sensors, etc. Micropump is among the imperative devices of fluid transport for biomedical applications. Micropump provides the actuation mechanism to deliver the particular amount of fluids (drug, blood, etc) through the fluidic system. The work on the development of micropump was initiated in 1970s. In 1980s, the microfabrication technology was primarily used for the development of micropumps. While in 1990s, the earliest microelectromechanical system (MEMS) based micropump was developed. Generally, MEMS based micropump has some important parts like chamber partition, diaphragm membrane, actuator, valves, channels, inlet, outlet, etc [2]. It is very important to understand the design specifications and parameters of these parts to develop suitable designs of micropump for device applications. There are two main types of micropumps: mechanical micropump with the mechanical moving part and non-mechanical micropump with no moving mechanical part. Mechanical micropumps can be further classifies into piezoelectric, thermopnuematic, shape memory alloy (SMA), electrostatic, phase change, bimetallic, electromagnetic and ion conductive polymer films (ICPF). Non-mechanical micropumps can be classified into electroosmotic, electrowetting, electrochemical, evaporation type, bubble type, magnetohydrodynamic, flexular planar wave (FPW) and electrohydrodynamic. All these kinds of micropumps have their own advantages and disadvantages. For the applications of microfluidic systems, the piezoelectric and electromagnetic micropumps have been presented by many researchers in literature. Although the fabrication process of piezoelectric actuator using piezo materials is complex on a single chip and piezopump needs high working voltage. However, piezoelectric micropump presents very fast response time and high actuation. So these are very useful for microfluidic applications. The structure of piezopump consists of piezoelectric material coating that is deposited on a film (membrane). Due to the induced voltage a definite measure of deformation occurs that acts as a pushing plate. This pushing plate drives out the fluid from the pump chamber. The schematic of a single chamber piezoelectric micropump is shown in the Fig. 1.

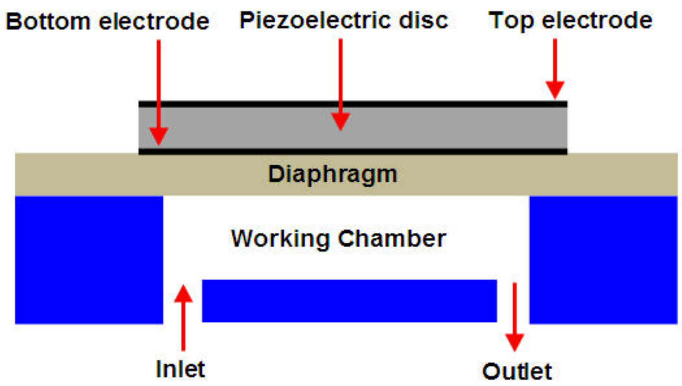

Fig. 1. Schematic of single chamber piezoelectric micropump.

Van Lintel et al. [3] fabricated the earliest piezoelectric micropump consisted of a chamber, passive check valve made of silicon, thin glass film actuated by piezodisk. On applying 125 V and I Hz, the maximum flow rate of 8 μL/min and back pressure of 9.8 kPa were obtained. Esashi et al. [4] presented a piezoelectric pump consisted of three layers. On applying 90 V and 30 Hz frequency, the pump showed a flow rate of 15 μL/min and back pressure of 6.4 kPa. Olsson et al. [5] presented a piezoelectric micropump consisting of two chambers with high performance. Mu et al. [6] studied the design optimization of piezoelectric micropump using different materials for actuator membrane and performed the numerical analysis. Cao et al. [7] performed the numerical modeling of piezoelectric actuator to envisage the membrane deflection. Koch et al. [8] reported a piezoelectric micropump and observed a flow rate of 120 μL/min and back pressure 2 kPa on applying voltage of 600 V and frequency of 200 Hz. Schabmueller et al. [9] presented a piezoelectric micropump consisting of passive valves. Ethanol was used in experiment and the pump showed the flow rate of 1500 μL/min and back pressure of 1 kPa. Fen and Kim [10] presented a piezoelectric micropump with single mode parylene valves and observed a flow rate of 3.2 μL/min and back pressure of 0.2 kPa at applied 80 V with less power consumption of 3mW. Geipel et al [11] presented a micropump with back pressure independency up to 20 kPa at low frequency. Trenkle et al. [12] presented a peristaltic micropump with piezostack actuation. Water was used in experiment and the pump showed a flow rate of 40 μl/min at the frequency of 28.6 Hz. Johari et al. [13, 14] fabricated a piezoelectric micropump for drug delivery system using two optical masks and with nano liter/minute flow rate. CoventorWare software was used for the fluidic analysis. Wang et al. [15] used a pumping system with a piezoelectric buzzer embedded in a channel to study the longitudinal flow asymmetry on pumping potential. Ali et al. [16] observed the quantitative dimensions of the pressure produced, AC frequency, applied electrical field and actuator length for the dynamic piezoelectric pump. Liu et al. [17] presented a high performance piezoelectric micropump consisted of four serial connected chambers in for insulin therapy system. Zhang et al. [18] reported a piezoelectric micropump with saw tooth shape microchannels driven by a travelling wave consisting of low resistance microchannel. Pham and Goo [19] reported a peristaltic micropump with three cylinder chambers, microchannels and two active membrane valves.

In this work, a novel design of hemofiltration (HF) system has been presented. The proposed system has various parts. Present work focuses on the design optimization and simulation of piezoelectric valveless blood pump for HF system. Valveless micropumps have open flow structure that makes them suitable for pumping of the particles, fluids and sensitive materials for use in biomedical applications. In addition, the design and fabrication of valveless micropumps is not complicated due to absence of moving mechanical parts like check valves/active valves. HF is the process of filtering the blood of patient suffering from acute renal failure. The waste products and water present in the patient's blood are filtered out using nano-porous hemofilter [20]. As excessive amount of plasma water is removed during HF process so there is a need to add the replacement fluid before inserting the filtered blood into the patient's body. The conventional devices or methods available for the filtration of blood for the patients with kidney failure have many disadvantages like inaccurate control, long term treatment process, repetitive insertions of needles, absence of biocompatibility, etc. That's why there is a promising need for development of a controlled and efficient blood filtration system.

2. Description of Hemofiltration (HF) System

The proposed MEMS based HF system consists of four modules that are electronic module, blood transport module, blood filtration module and drug delivery module. The electronic module supplies the essential functionalities for proper system working. Microneedles have been used for the extraction of blood from patient body and then the blood is pumped through the filtration module using piezoelectricaly actuated blood pump. The blood filtration module consists of hemofilter, mixing chamber, heparin pump, heparin reservoir, drain bag, air trap detector and replacement fluid reservoir. Heparin is mixed with the blood in mixing chamber. Hemofilter with hexagonal nano pores removes the smaller molecular weight toxins having molecular weight < 500 like urea, potassium, phosphate, creatinine, uric acid, etc; higher molecular weight toxins having molecular weight > 500 and <40,000 like inulin, β2 microglobulin, inflammatory mediators, cystatin, etc and some protein bound toxins like methylglyoxal, indoxyl sulfate, leptin, etc. While the blood cells, protein and other large molecules remain in the blood of patient as there molecular weights are too high to pass through the filter. The replacement fluid is inserted into the blood before entering into body to overcome the removal of plasma water during HF process. Then the filtered blood is inserted into the patient body through microneedles (blood transport module). Drug delivery module provides the drugs during the filtration if there is an urgent need of drug. The schematic of proposed hemofiltration system is shown in Fig. 2 [20].

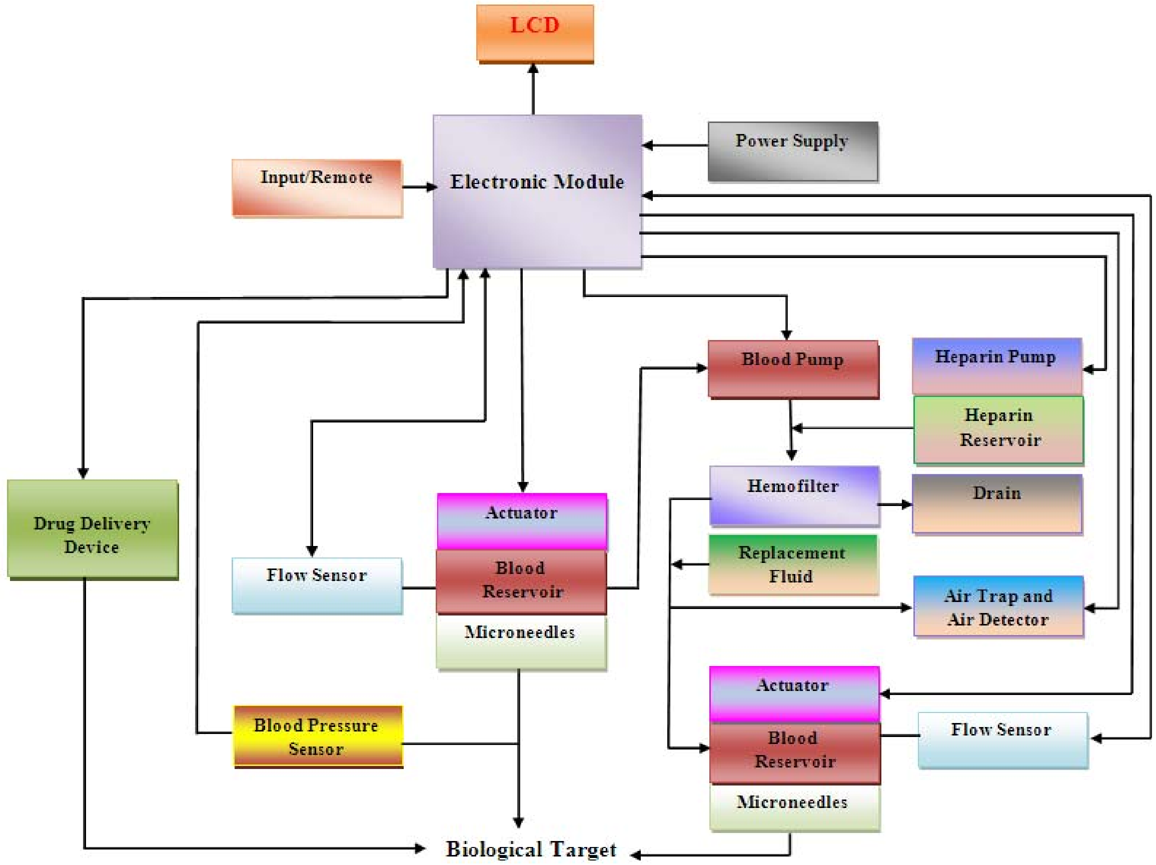

Fig. 2. Schematic of MEMS based hemofiltration system.

3. Micropump Design and Theoretical Analysis

Design of blood micropump is extremely important in HF system to achieve the desired rate of fluid flow in HF process. Polymeric valveless micropump with three actuators has been designed for HF system. A schematic design of valveless micropump is shown in Fig. 3. Micropumps with nozzle/diffuser elements utilize the different pressure drop characteristics. In supply mode, the membrane of actuator moves upward and volume of chamber increases. Due to that motion, more fluid enters through inlet than outlet. While in pump mode, actuator moves downward and chamber volume decreases. Hence more fluid flows through the outlet than inlet.

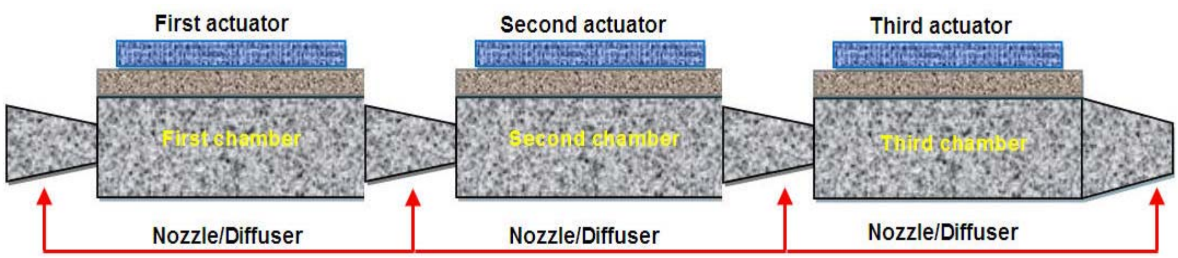

Fig. 3. Schematic of three chamber micropump.

Micropump actuator consists of polydimethylsiloxane (PDMS) membrane and PZT-4 disc attached together. The actuator is bonded with microchannels and nozzle/diffuses elements to form blood micropump. The dimensional model of piezoelectric valveless micropump is shown in Fig. 4.

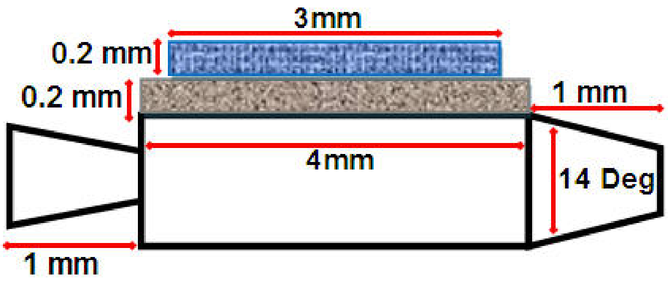

Fig. 4. Model dimension of blood micropump.

Piezoelectric analysis involves coupling of structural and electric fields. On applying voltage, the piezoelectric material shows bending that result into the generation of voltage due to vibration process. In piezoelectric actuator, there is present a piezo disk bonded with membrane of PDMS. According to the piezoelectricity theory, the stress T_r^P around the perimeter of piezoelectric actuator can be defined as given in equation (1) [21]:

$$T_r^P = -\frac{ed'V_0}{(1-\sigma)h}, \tag{1}$$

where, e is the electric field, V_o is the applied voltage, h is the thickness of actuator and d' is the piezoelectric coefficient of coupling effect.

The simultaneous inclusive coupling related to piezoelectric actuators can be written in the matrix form as given in equation (2) [21]:

$$\begin{bmatrix} [k_{11}][k_{12}] \\ [k_{21}][k_{22}] \end{bmatrix} \begin{Bmatrix} \{X_1\} \\ \{X_2\} \end{Bmatrix} = \begin{Bmatrix} \{F_1\} \\ \{F_2\} \end{Bmatrix}, \tag{2}$$

where, $[k_{i,j}]$; i, j = 1, 2 are sub-matrices of; $[F_i]$; i= 1, 2 are vectors for force constraint; $[X_i]$; i, j = 1, 2 are degree of freedom (DOF) types.

In linear piezoelectric analysis, elasticity constants are joined with the charge electrostatics constants using the constants of piezoelectricity as given in equation (3) (ANSI/IEEE Std 176 (1987)) [22]:

$$\begin{Bmatrix} [T] \\ [D] \end{Bmatrix} = \begin{bmatrix} [c]^E\ [e] \\ [e]^\tau\ [\varepsilon] \end{bmatrix} \begin{Bmatrix} [S] \\ [-E] \end{Bmatrix} \tag{3}$$

where, $[T]$ is the vector for stress constraint, $[D]$ is the vector for electric flux density, $[S]$ is the vector for strain, $[E]$ is the vector for electric field strength, $[c]^E$ is the matrix for elasticity at constant values of E, $[e]$ is the matrix for piezoelectric stress constraint, $[e]^T$ is the matrix for dielectric at unvarying values of strain.

The governing equation for incompressible flow of fluid in piezoelectric micropump with valveless structure can be given as in equations (4, 5):

$$\rho_L \frac{dv}{dt} = \rho_L g + \mu \nabla^2 V - \nabla P \tag{4}$$

$$\frac{\partial \rho_L}{dt} + (V.\nabla)\rho_L = 0 \tag{5}$$

where, V is the vector for velocity; μ is the fluid viscosity; P is the dynamic pressure; L_ρ is the liquid density.

For supply mode, the inlet works like the diffuser and the outlet works like the nozzle as the chamber volume increases. So, the fluidic flow through the inlet part Q_{in} and the outlet part Q_{out} can be given as in equations (6, 7) [23]:

$$Q_{in} = Q_d = A_1\left(\frac{2}{\rho}\right)^{1/2}\left(\frac{\Delta p_d}{k_d}\right)^{1/2} \tag{6}$$

$$Q_{out} = Q_n = A_1\left(\frac{2}{\rho}\right)^{1/2}\left(\frac{\Delta p_n}{k_n}\right)^{1/2} \tag{7}$$

For pumping mode, the inlet works like the nozzle and the outlet works like the diffuser as the chamber volume decreases. So, the fluidic flow through the inlet part Q_{in} and the outlet part Q_{out} can be given as in equations (8, 9) [23]:

$$Q_{in} = -Q_n = A_1\left(\frac{2}{\rho}\right)^{1/2}\left(\frac{\Delta p_n}{k_n}\right)^{1/2} \tag{8}$$

$$Q_{in} = Q_d = A_1\left(\frac{2}{\rho}\right)^{1/2}\left(\frac{\Delta p_d}{k_d}\right)^{1/2} \tag{9}$$

The coefficients k_d and k_n for pressure loss through the directions of diffuser and nozzle elements can be written as given in equations (10, 11) [24]:

$$k_d = K_{d,en} + K_{d,l} + K_{d,ex}\left(\frac{A_1}{A_2}\right)^2 \tag{10}$$

$$k_n = K_{n,en} + (K_{n,l} + K_{n,en})\left(\frac{A_1}{A_2}\right)^2, \tag{11}$$

where $K_{d,en}$, $K_{d,ex}$ are the coefficients of pressure loss for the diffuser element at inlet and outlet positions. $K_{d,l}$, $K_{n,l}$ are the length coefficients around the perimeter of diffuser and nozzle element. $K_{n,en}$, $K_{n,ex}$ are the coefficients of pressure loss for the diffuser element at inlet and outlet positions. A_1, A_2 are the smallest and largest areas of the diffuser and nozzle element respectively.

The diffuser efficiency η can be written as the ratio of pressure loss coefficient k_n for nozzle to pressure loss coefficient k_d for diffuser as equation (12).

$$\eta = \frac{k_n}{k_d} \tag{12}$$

4. Numerical Simulation

The numerical simulation of proposed micropump has been performed in ANSYS for design optimization of the piezoelectricaly actuated valveless blood pump for HF system. The multiple code coupling scheme has been used in which the ANSYS-structural model couples the information with CFX-fluid model and vice versa. Forces, displacement and other parameters are coupled for information/data transfer. Three dimensional (3D) model of the blood micropump with three actuators has been built for multifield analysis. The pump model with boundary conditions in ANSYS-Multiphysics and ANSYS-CFX has been shown in Fig. 5.

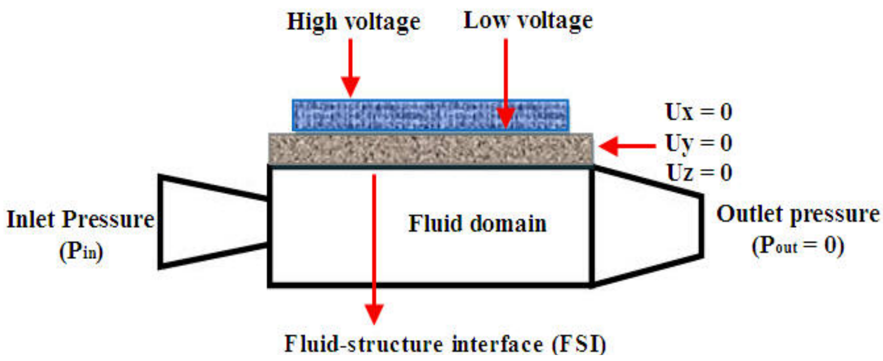

Fig. 5. Model dimension of blood pump.

Two different mechanical inputs have been modeled for actuator and fluidic model. PDMS membrane has been attached with PZT-4 disc to form actuator. Serial microchannels have been bonded with the actuators to form micropump. The material properties of PZT-4 and PDMS are given in Table 1.

SOLID-5 element has been used for displacement and voltage degrees of freedom. SOLID-186 element has been used for actuator membrane. Blood and heparin with 2:1 have been used as working fluid for multifield simulation. Both mechanical inputs have been solved in ANSYS launcher. Coupled-multifield results have been acquired for ANSYS and CFX codes using MFX-ANSYS/CFX environment. The fluid CFX model of blood pump is shown in Fig. 6.

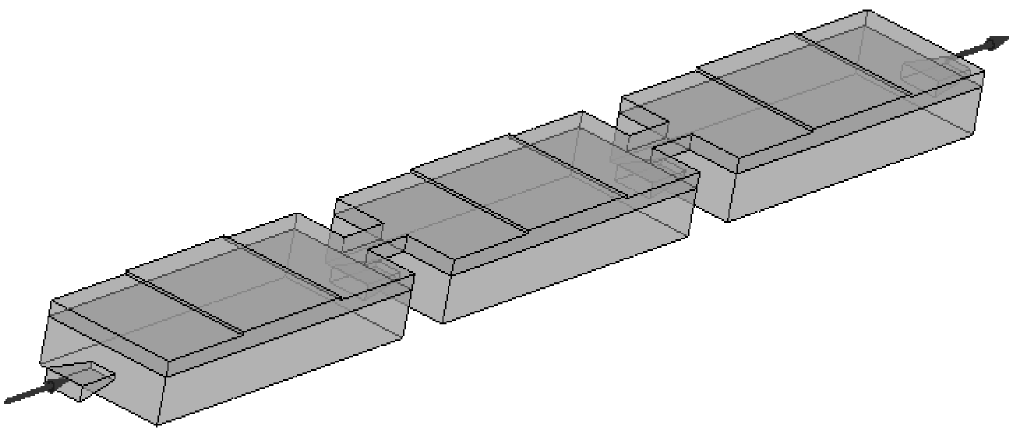

Fig. 6. CFX model of micropump.

Table 1. Material properties of PZT-4 and PDMS.

Material	Properties	Values	
Piezoelectric (PZT-4)	Density (Kg/m^3)	7500	
	Elasticity stiffness matrix (N/m^2)	$\begin{bmatrix} 13.2 & 7.3 & 7.1 & 0 & 0 & 0 \\ 7.3 & 11.5 & 7.3 & 0 & 0 & 0 \\ 7.1 & 7.3 & 13.2 & 0 & 0 & 0 \\ 0 & 0 & 0 & 2.6 & 0 & 0 \\ 0 & 0 & 0 & 0 & 2.6 & 0 \\ 0 & 0 & 0 & 0 & 0 & 3 \end{bmatrix} 10^{10}$	
	Piezoelectric stress matrix (C/m^2)	$\begin{bmatrix} 0 & -4.1 & 0 \\ 0 & 14.1 & 0 \\ 0 & -4.1 & 0 \\ 10.5 & 0 & 0 \\ 0 & 0 & 10.5 \\ 0 & 0 & 0 \end{bmatrix}$	
	Permittivity	x	804.6
		y	659.7
		z	804.6
PDMS	Young's modulus (N/m^2)	0.003	
	Density (Kg/m^3)	965	
	Poisson's ratio	0.4	

The maximum displacement in actuator occurs at the centre that is shown in Fig. 7. The maximum deflection 6.808 μm is obtained at 105 V with frequency of 250 Hz in each actuator.

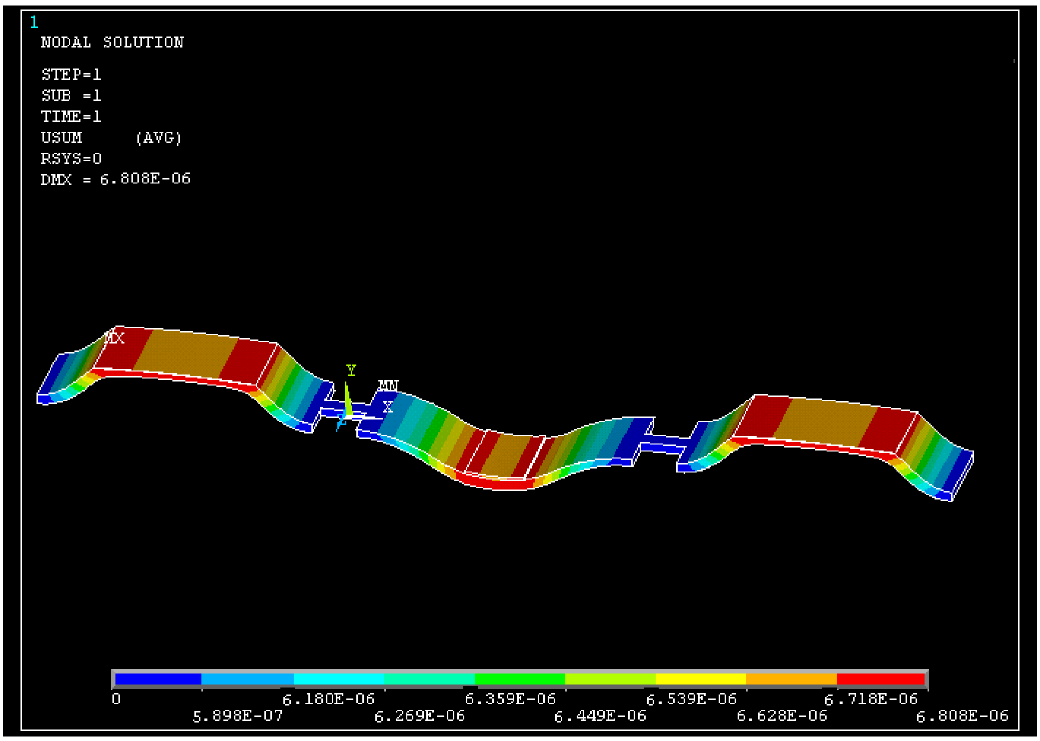

Fig. 7. Deflection of serial actuators at applied 105 V.

In vector form the deflection in serial actuators is shown in Fig. 8. First and third actuators move in same directions (upward and downward) while the central actuator deflects in opposite direction.

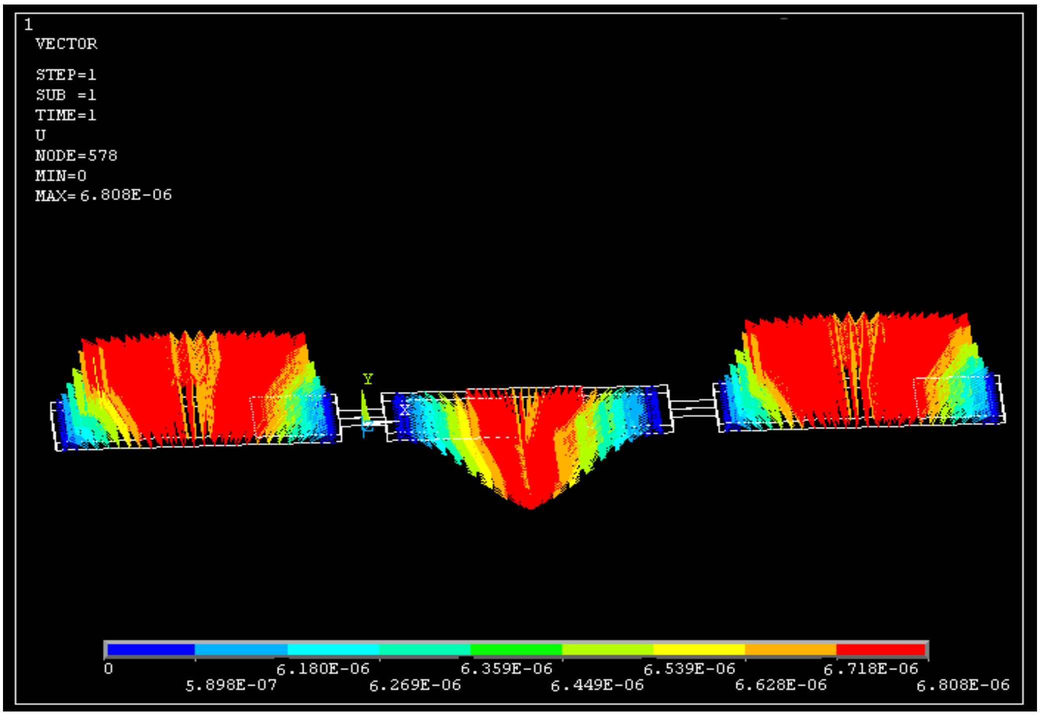

Fig. 8. Vector form of serial actuators deflection at applied 105 V.

It has been observed that maximum von mises stress in actuator occurs near the interface between piezoelectric disc and membrane that is shown in Fig. 9. The resultant stress values due to applied voltage are well below the yield strength of the materials.

Fig. 9. Stress distribution in the actuators.

Due to stress in the actuators, the mechanical strain occurs in the structure of serial actuators that is shown in Fig. 10.

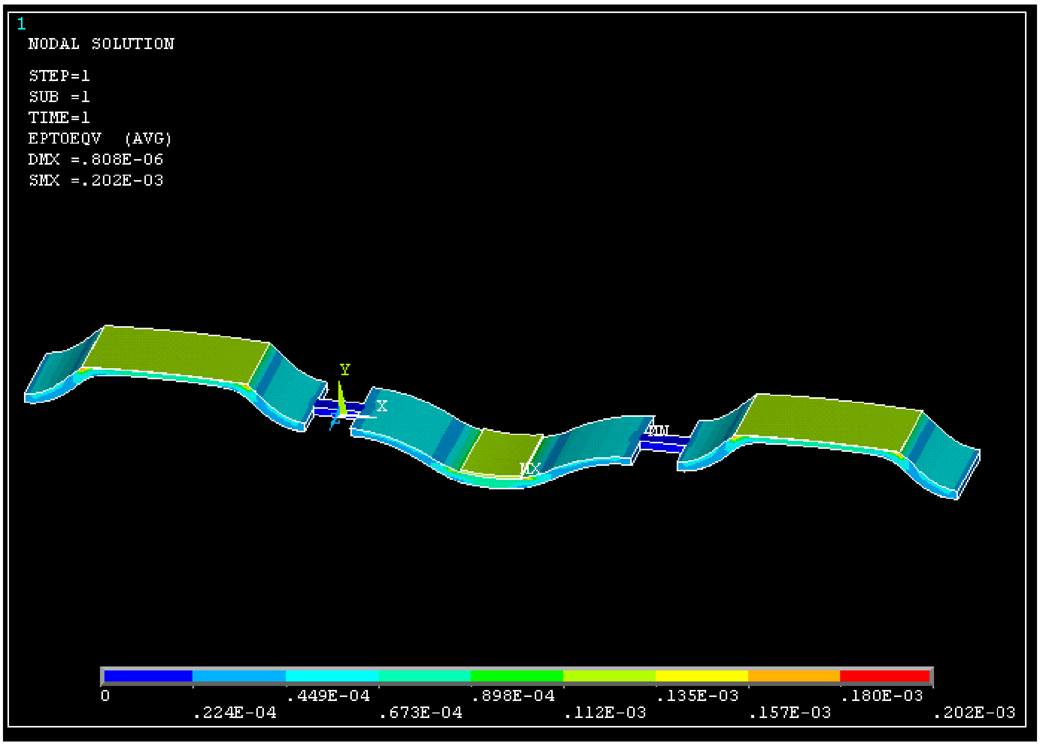

Fig. 10. Mechanical strain in the serial actuators.

5. Results and Discussion

The deflection of piezoelectric actuator along the length of actuator membrane has been studied as shown in Fig. 11. The maximum deflection occurs at the center of actuator. Results show that the curve of simulated deflection of actuator is in close agreement with the curve of theoretical deflection. That proves the validity of the analysis performed using code coupling in ANSYS and CFX solvers.

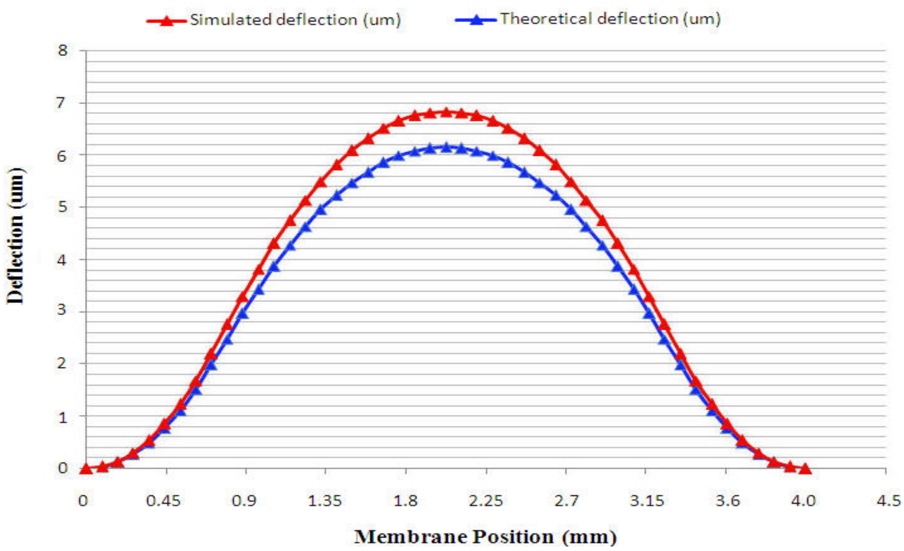

Fig. 11. Deflection of PDMS membrane at various position.

When voltage is applied on piezo actuator, it deforms perodicalyy up and downward direction due to piezoelectric effect. The maximum deflection of 6.808 μm occurs at applied voltage of 105 V as shown in Fig. 12.

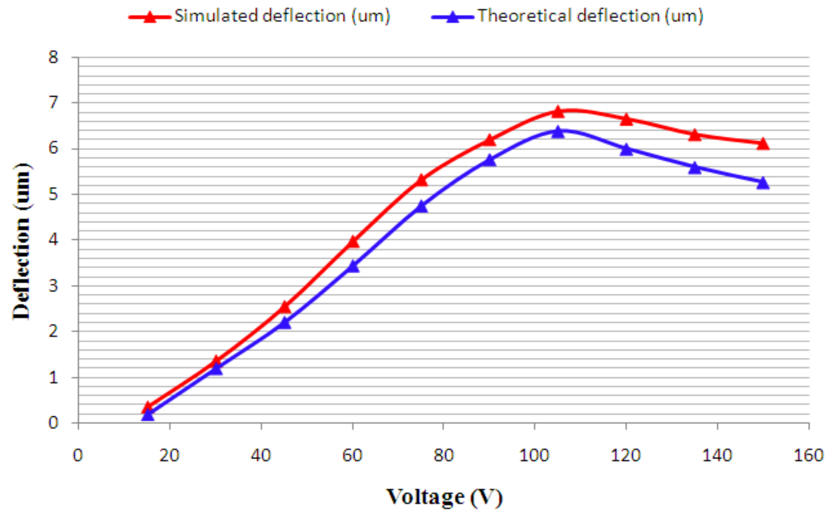

Fig. 12. Relation between voltage and actuator deflection.

Membarne thickness impose directly impact on deflection at constant voltage. That causes the change in flow rate as shown in Fig. 13.

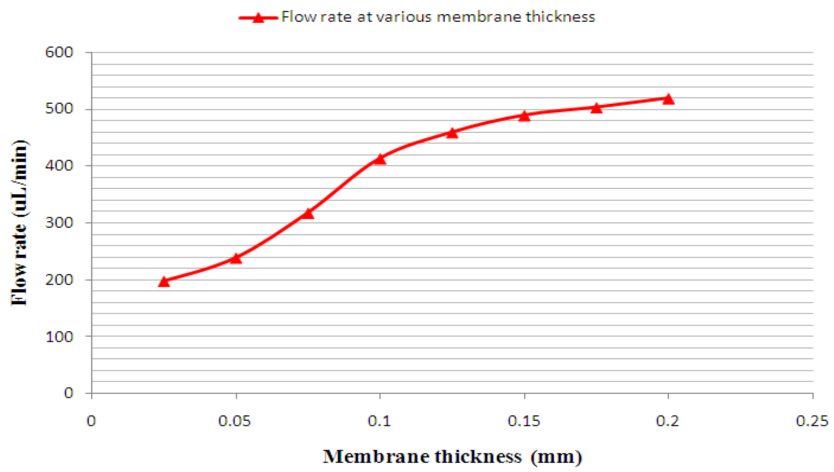

Fig. 13. Effect of PDMS membrane thickness on flow rate.

In supply and pump mode the impact of diffuser angle on flow rate is shown in Figs. 14 and 15 respectively. In summply mode, more fluid enters in inlet than outlet at applied 105 V.

Efficiency of diffuser increases by increasing diffuser angle at different piezoelectric disc thickness by keeping the thickness of PDMS membrane constant as shows in Fig. 16.

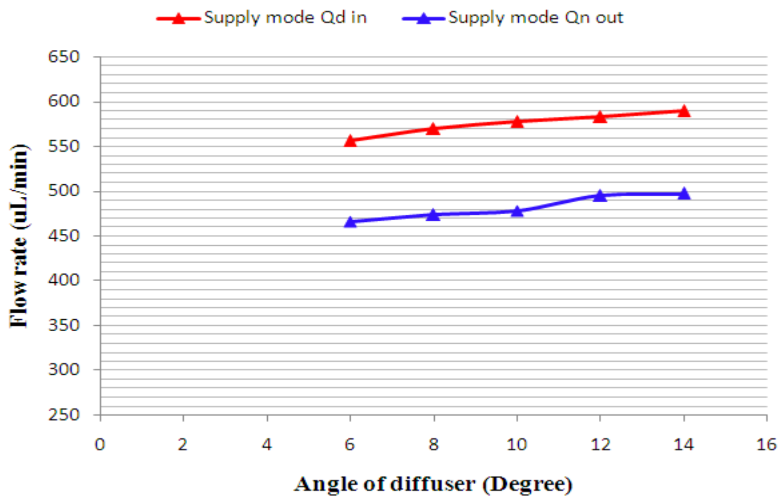

Fig. 14. Relation between diffuser angle and flow rate in supply mode.

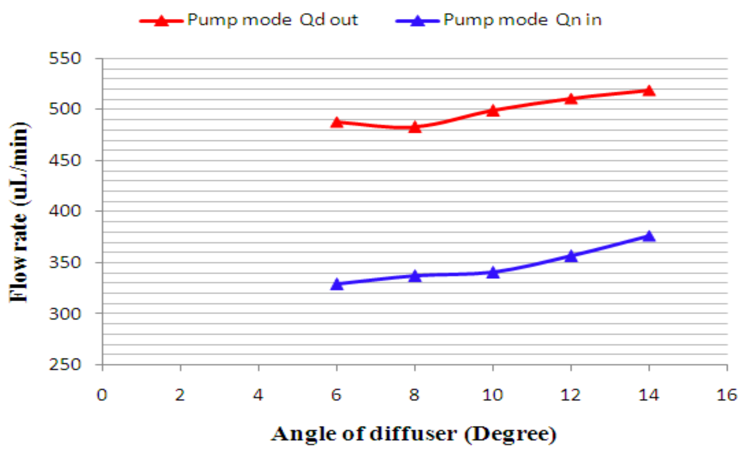

Fig. 15. Relation between diffuser angle and flow rate in pump mode.

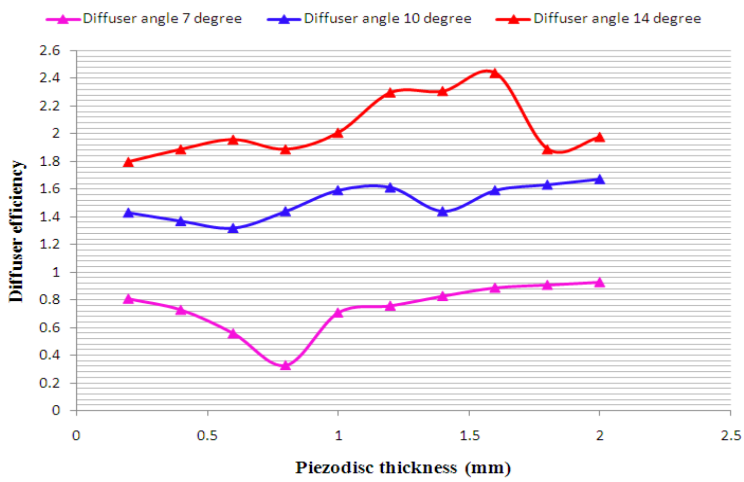

Fig. 16. Effect of diffuser angle on diffuser efficiency.

Variation in flow rate at 60 V, 105 V and 150 V with different applied frequencies is shown in Fig. 17. There is a small effect on deflection of actuator by varying the frequency. That's why flow rate slowly increases with the variation in frequency. Maximum flow rate of 519 µL/min has been obtained at 200 Hz with applied 105 V.

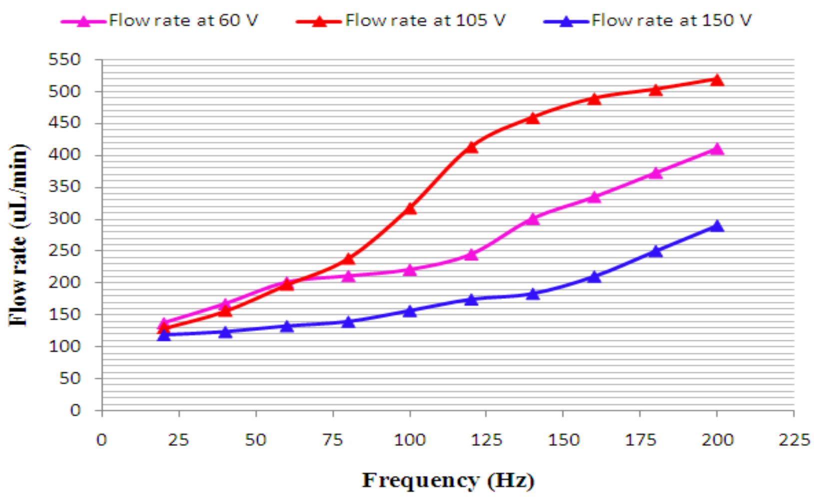

Fig. 17. Effect of varying frequency on flow rate at different voltage values.

The relation between applied voltage and flow rate of piezoelectric micropump has been shown in Fig. 18. It has been observed that the flow rate increases with increase of applied voltage up to 110 V and then it decreases with further increase in voltage at same frequency. The maximum flow rate of 519 µL/min has been obtained at 105 V. Simulated results of flow rate at varying voltage values are in close agreement with the theoretically calculated results.

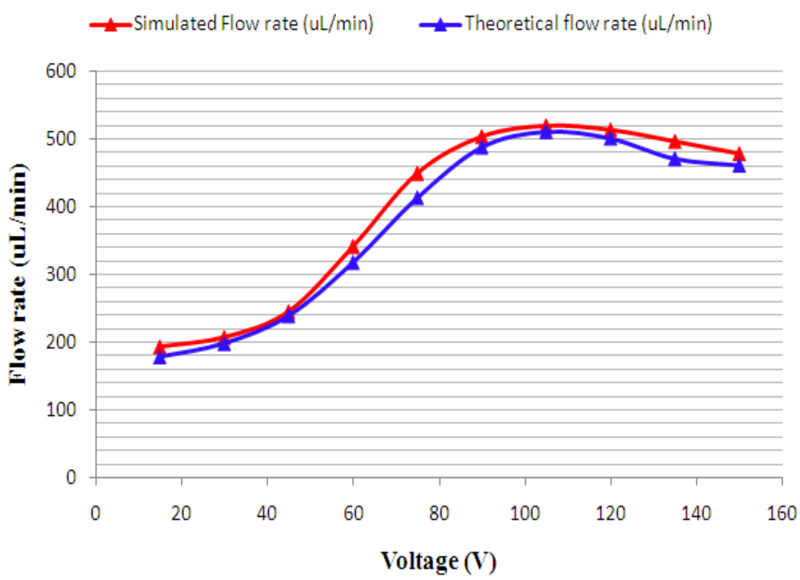

Fig. 18. Effect of applied voltage on flow rate.

The effect of back pressure on flow rate of piezoelectricaly actuated valveless micropump with serial actuators has been shown in Fig. 19. As the back pressure increases, the flow rate of micropump decreases. Results shows that a maximum backpressure of 2 .5 kPa occurs at applied 105 V.

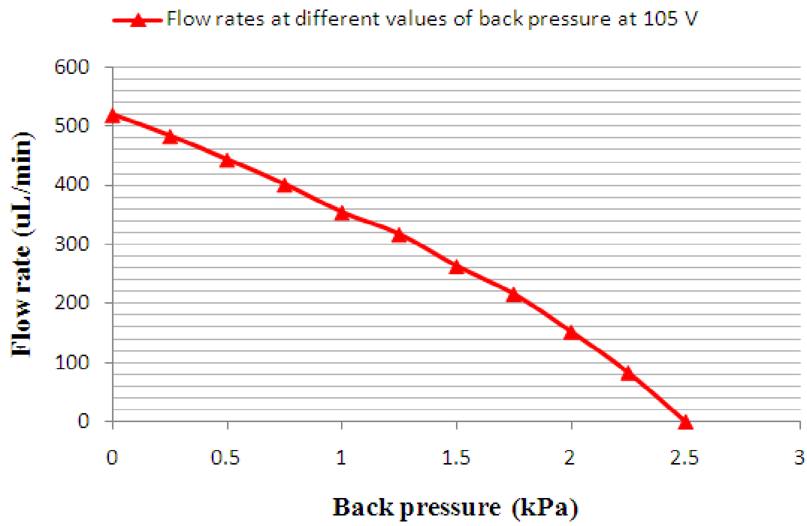

Fig. 19. Effect of backpressure on flow rate at 105 V.

Fig. 20 shows the power consumption of the piezoelectric valveless micropump with respect to the flow rate at 105 V. Power consumption is increases with the increase in flow rate. It is observed that the maximum power consumption of 45 mW has been obtained for the micropump at highest flow rate of 519 µL/min.

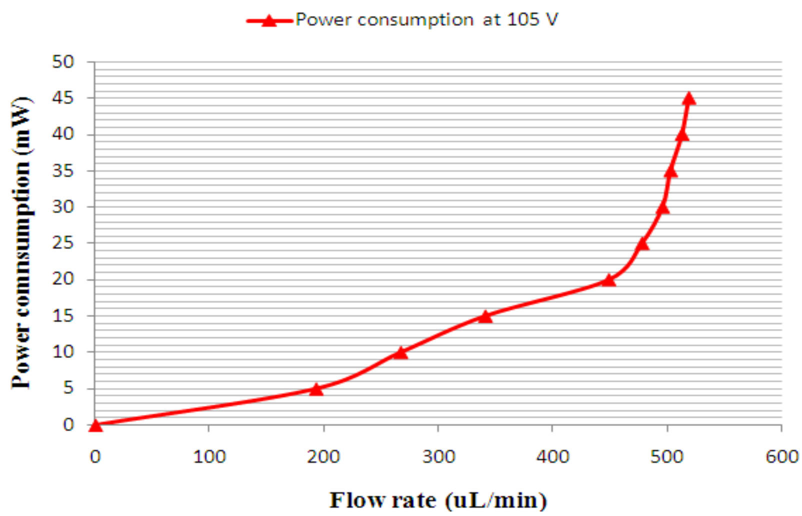

Fig. 20. Relation between flow rate and power consumption.

Fig. 21 describes the variation in power consumption at different values of driven frequency at 105 V. The maximum power consumption was obtained at a driven frequency of 200 Hz. This is the point when the proposed micropump produces its maximum exterior work and the highest value of power consumption was achieved due to the conservation of energy.

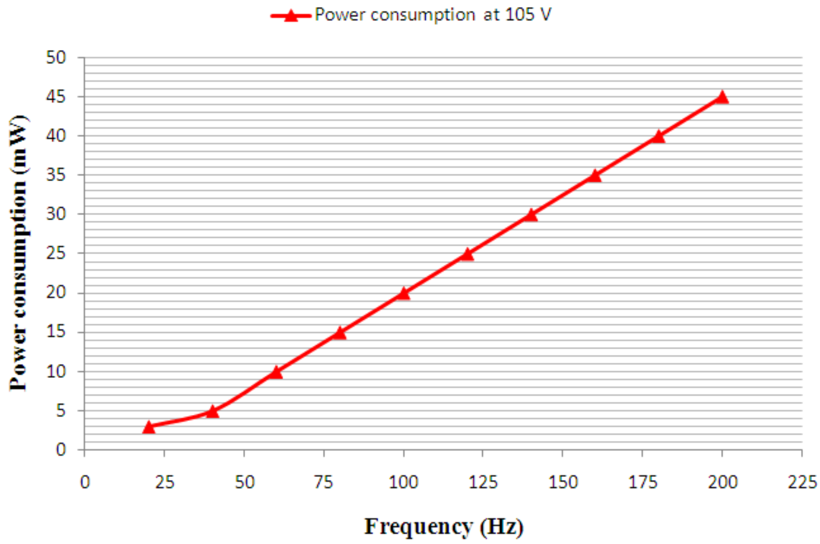

Fig. 21. Relation between frequency and power consumption.

6. Conclusions

Kidney failure is a very common disease and increasing rapidly every day. The conventional method of blood filtration are tiresome, throbbing and painful for the patients because of frequent needle inoculations, long term treatment process, lack of biocompatibility and side effects. Thus, there is a need to develop a low cost and high performance blood filtration system for patients with kidney diseases who require to remove the waste products from their blood to save their lives. Here, we present a novel design of MEMS based hemofiltration (HF) system. The proposed HF system consists of four modules that are electronic module, blood transport module, blood filtration module and drug delivery module. Coupled multifield analysis of MEMS based piezoelectrically actuated polymeric valveless micropump has been performed for HF system. Micropump comprises of thee serial actuators, nozzle/diffuser elements and three chambers. A numerical technique has been adopted for design optimization of piezoelectric valveless blood micropump before fabrication because of the complexities linked with modeling of piezoelectric micropump. Effects of various parameters such as actuator thickness, frequency, voltage, nozzle/diffuser angle, power consumption on flow rate have been investigated during the multifield analysis. The maximum deflection of 6.808 μm has been observed at applied voltage of 105 V. Maximum flow rate of 519 μL/min has been obtained at 200 Hz with applied 105 V by keeping constant membrane thickness of 0.2 mm. The proposed micropump would present valuable information and predictable data for MEMS based HF system.

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Guide for Contributors

Aims and Scope

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Contributions are invited on all aspects of research, development and application of the science and technology of sensors, transducers and sensor instrumentations. Topics include, but are not restricted to:

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- Smart sensors and systems;
- Sensor instrumentation;
- Virtual instruments;
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