Numerical Simulation of Piezoelecrtic Diffuser Nozzle Valve Micropump

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Abstract—**In the case of low flow rate applications like drug delivery systems, cooling of tiny electrical elements require special pump which provide very low flow rate. Piezoelectric single chamber diffuser nozzle type valveless pump is an attractive device to be used as a micro pump for low flow rates. Here the pump converts the reciprocating motion of a diaphragm, activated by a piezoelectric disk, into a pumping action. Instead of conventional valves, which have moving parts, nozzle/diffuser elements that have a preferential flow direction are used to direct the flow from the inlet to the outlet. A mathematical model that can simulate the pump performance under given geometrical and operational conditions is essential for the optimal design of such pumps. The proposed course of work is about Numerical simulation of mathematical model of such a pump.For this purpose the feasibility of various software such as MATlab simulation and ANSYS/COMSOL are used.**

Index Terms— **valve less pump, piezoelectric ,micropump**

I. INTRODUCTION

In the case of medical application, space exploration and micro electric cooling, the rate of flow should be low in the order of millimetre. The small fluid volumes in these systems are often pumped, controlled or otherwise manipulated during operation. So in this introduction, we consider a few applications briefly to gain insight in to design relevant to micropumps[1].

Among the first micropumps, those developed by Jan Smits in the early 1980s were intended for use in controlled insulin delivery systems for maintaining diabetics' blood sugar levels without frequent needle injections. High volumetric flow rates are not likely to be required of implanted micropumps. Reciprocating displacement micropumps, the most widely reported micropumps, have been produced with a wide variety of chamber configuration, valve types, drivers and constructions. Piezoelectrically driven reciprocating displacement micropumps have been the subject of particular attention and are now available commercially.

When we are going to analyse different types of micropumps, mainly two types reciprocating and dynamic micropumps[6]. As we are trying to reduce the size of the micropump or trying to get low flow rate centrifugal type cannot become more efficient compared with reciprocating type. So centrifugal micropump is inefficient in low Reynolds number flow application [2].Thus concluded that reciprocating type micro pump is more efficient that dynamic micropump in the case of low flow rate. In reciprocating type there are different combination of drivers, valves and chamber type. Study of these different configurations will lead to the best combination of aforementioned features

In the literature review we are analysed different types micropump and the configuration. Reciprocating type with piezoelectric driver pumps have more advantages than other types of reciprocating type. Since the centrifugal type micropump has limitations in reducing size and moving parts, it has less application in microelectrical mechanical systems. Single chamber chamber configuration has more advantages than other type chamber configuration and in the case of valve, diffuser/nozzle is more efficient since it does not have any moving part. Thus we can concluded that piezoelecrtic single chamber diffuser nozzle type valve reciprocating micropump have high performance than other type[1][2][6][7].

Many of work followed the above discussed features of micropump. Consider one of the work which follow the aforementioned features of micropump. In that they are discussing the modeling and characterization of a MEMS micropump with circular bossed membrane [4].

The objective of proposed course work is about Numerical simulation of mathematical model of Piezoelectric single chamber diffuser/nozzle type pump. For this purpose the feasibility of various software such as MAT lab simulation and ANSYS/COMSOL are used.

II. METHODOLOGY

Modelling of microfluidic systems in general and micropumps specifically involves elastic mechanics, fluid transmission, and energy transformation among electrical, magnetic, and thermal fields, as well as energy combined effects. The equations governing the behaviour of these systems often have no analytical solutions, particularly if they include nonlinear effects such as large deflection and stress stiffening. One method is to develop a differential equation which describe the micropump's behaviour after all individual elements have been functionally constructed. The key components of a micro-electro-mechanical system (MEMS). micropump include membrane, membrane pillar , microchamber, inlet/outlet micro-valves, and inlet/outlet microchannels (Fig 1). The membrane pillar is used to generate large drawing and pumping force.The membrane is in its neutral, or flat, position when the external pressure and the pressure inside the micro chamber are balanced. When it is actuated by electrosataically, the membrane deforms downward, the outlet microvalve is opened (the inlet microvalves will be closed instantaneously), and the microchamber volume is reduced. That condition called "Pump Mode". On the other hand, while the membrane is driven upward, the fluid that contains drugs is drawn into the microchamber via the inlet microvalves through the inlet microchannels (outlet microvalve is closed). This is identified as the "supply mode."For the fabrication of micropump etching process is normally carried out especially silicon etching[5].

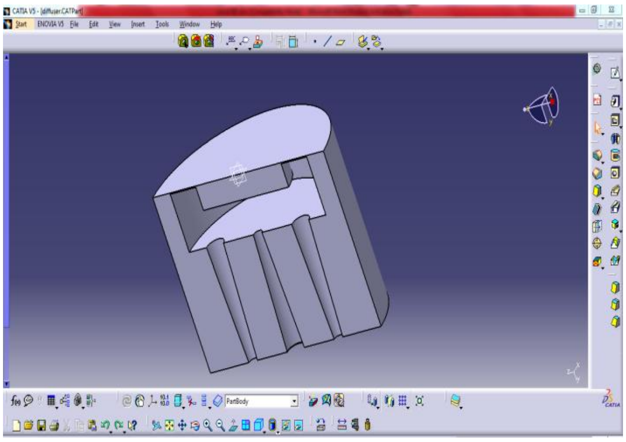

Figure 1. Schematic diagram of micropump

A. Governing equation of the model

For full system modelling and simulating one method is to develop differential equation to describe the micropump behaviour. The continuity equation, which connects the mass flow entering/exiting the microchamber with transient fluid mass change in the microchamber, follows

$$\oiint_{cs} (n \cdot \rho v)\, dA = -\frac{\partial}{\partial t}\iiint_{cv} \rho\, d\forall \qquad (1)$$

The left side of Eq (1) is equivalent to the net efflux of fluid driven out of the microchamber, and is determined by the characteristics of inlet/outlet passive check microvalves. If the inertia of valves and the acceleration of fluid are small and negligible, its quasistatic ODE is

$$\oiint_{cs} (n \cdot \rho v)\, dA = \rho(\phi_{0v}(p_{ov}) - \phi_{iv}(p_{iv}) \qquad (2)$$

Where Φov and Φiv are the steady-state (static) flow rates of outlet and inlet microvalves as functions of Pov and Piv, respectively. Pov and Piv are the hydrostatic pressure difference across the outlet and inlet microvalves as functions of time (t). They can be expressed as

$$P_{ov}(t) = P_{\mathrm{ch}}(\mathrm{t}) - P_{\mathrm{out}}$$

$$P_{iv}(t) = P_{\mathrm{in}} - P_{\mathrm{ch}}(\mathrm{t})$$

Consider the RHS of equation (1) which is equivalent to rate of change of mass within the control volume, assuming that pumping medium is incompressible.

$$-\frac{\partial}{\partial t}\iiint_{cv} \rho\, d\forall = -\left(\rho\frac{\partial V_{ch}}{\partial t} + \rho\frac{\partial V_o}{\partial t} - \rho\frac{\partial V_{iv}}{\partial t} + \rho \mathrm{c} - \frac{\partial(\rho_{gas}V_{gas})}{\partial t}\right) \qquad (3)$$

Equating Eqs (2) and (3), and rearranging the terms, we get

$$\frac{\partial V_{ch}}{\partial t} = (\phi_{iv}(P_{ov}) - \phi_{iv}(P_{iv}) - \frac{\partial V_o}{\partial t} + \frac{\partial V_o}{\partial t} - \frac{\partial V_{ov}}{\partial t} + \frac{1}{\rho}\frac{\partial(\rho_{gas}V_{gas})}{\partial t} \qquad (4)$$

Vch describes the coupling between the volume displacement of the membrane and the resulting pressure inside the microchamber which varies with the microchamber pressure (Pch) and driving pressure (P), and $\frac{\partial Vch}{\partial t}$is

$$\frac{\partial V_{ch}}{\partial t} = \frac{\partial V_{ch}}{\partial t}\bigg| P_{ch} + \frac{\partial V_{ch}}{\partial t}\bigg| P$$

$$= \frac{\partial V_{ch}}{\partial t}\bigg| P_{ch} . \frac{dP}{dt} + \frac{\partial V_{ch}}{\partial t}\bigg| P . \frac{dP_{ch}}{dt} \qquad (5)$$

Substituting the value of Eq(5) in Eq(4) and rearranging the term for rate of change of pressure inside the chamber.

$$\frac{dP_{ch}}{dt} = \frac{(\phi_{iv}(P_{ov}) - \phi_{iv}(P_{iv}) - \frac{\partial V_{ch}}{\partial P}\Big| P_{ch} . \frac{dP}{dt}}{\frac{\partial V_{ch}}{\partial P_{ch}}\Big| P + \frac{\partial V_o}{\partial P_{ch}} - \frac{\partial V_{iv}}{\partial P_{ch}} + \frac{\partial V_{ov}}{\partial P_{ch}} - \frac{1}{\rho}\frac{\partial(\rho_{gas}V_{gas})}{\partial t}} \qquad (6)$$

This is the governing equation for evaluating rate of change of pressure inside the micro chamber. From this equation it is clear that, pressure distribution inside the microchamber is time dependent variable and there is no variation with respect to position.

Since the wall of the chamber is made up with silicon, the side of the wall is considered as rigid. So rate of change of volume of microchamber with respect to chamber pressure is neglible. The volumetric deformations caused by microvalves are small compared to the stroke volume of the microchamber, hence volumetric displacement change of inlet and outlet valve can be ignored .In the case that the micropump is primed before running, the term (ρgasVgas) will be zero. then the Eq(6) reduced to

$$\frac{dP_{ch}}{dt} = \frac{(\phi_{iv}(P_{iv}) - \phi_{ov}(P_{ov}) - \frac{\partial V_{ch}}{\partial P}\Big| P_{ch} . \frac{dP}{dt}}{\frac{\partial V_{ch}}{\partial P_{ch}}\Big| P} \qquad (7)$$

The flow rate equation for inlet microvalves, Φiv ,is derived by fitting the curve. Because the flow rate in the backward flow is very small compared to that of the forward flow , it is assumed that the inlet microvalve has no leakage. The fourth-order fitted curve of Φiv as a function of Pch(t), for $0 \leq \Delta P \leq 10$ kPa is

$$\Phi iv(Piv) = (2.038e^{-5})Pch^4 + (3.56e^{-3})Pch^3 + (0.2456)Pch^2 - (0.389)Pch - 2.867 \quad (8)$$

And for Piv ≥10Kpa

$$\Phi iv(Piv) = (1.679e^{-5})Pch^4 + (5.727e^{-3})Pch^3 + (6.488e^{-2})Pch^2 - (0.2404)Pch + 2.338e^{-6} \quad (9)$$

The flow rate equation for outlet microvalves, Φov,is derived by the same methodolgy curve fitting. The curve-fitting equation of the ouletflow rate for $0 \leq \Delta P \leq 10$ kPa, is

$$\Phi ov(Pov) = (1.679e^{-5})Pch^4 - (5.727e^{-3})Pch^3 + (6.488e^{-2})Pch^2 - (0.2404)Pch + 2.338e^{-6} \quad (10)$$

And for Pov ≥10Kpa

$$\Phi ov(Pov) = (2.038e^{-5})Pch^4 - (3.56e^{-3})Pch^3 + (0.2456)Pch^2 - (0.389)Pch - 2.867 \quad (11)$$

The term dp/dt in the Eq (7) is the rate od change of driving pressure with respect to time .Assuming that a harmonic driving pressure is used

$$P(t)=P0[1-\cos(2\pi ft)] \quad (12)$$

where P0 is the amplitude and f is the driving frequency. P(t) should always be smaller than the maximum allowable net pressure, 80 kPa. The maximumP(t) occurs when [1-cos(2πft)]= -1. In this case, [1-cos(2πft)]equals 2; thus, Po,max = 40 kPa.

Substituting Eqs (7)-(12) into Eq (7), the rate of change of pressure inside the microchamber is

$$\frac{dP_{ch}}{dt} = 458.505\phi_{iv}(P_{iv}) - 29.253(\phi_{ov}(P_{ov}) + 2\pi f P0 \sin(2\pi f t) \quad (13)$$

The above Equation can be solved in MAT LAB.

IV. RESULTS AND DISCUSSIONS

A. *Pressure vs time*

In this logical diagram represent the combined effect of driving pressure and pressure inside the chamber along with time. For that input signal is tapped from the main line to the simulate the equation P(t)=P0[1-cos(2πft)] ,which connected to the MUX operator along with the output from the integrator .Thus we can see the combined effect of driving pressure and chamber pressure(Fig 2 and fig 3).

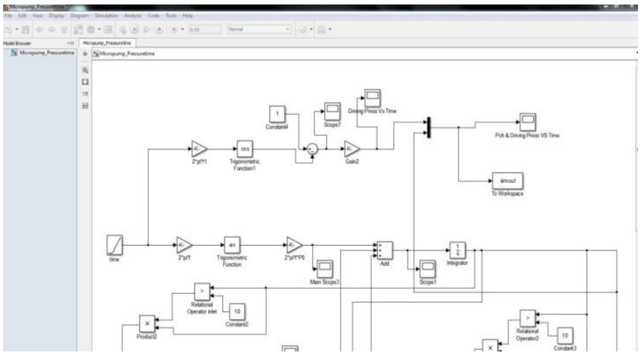

Fig 2 Microchamber_Pressure(Pch,Pdrive) Vs Time

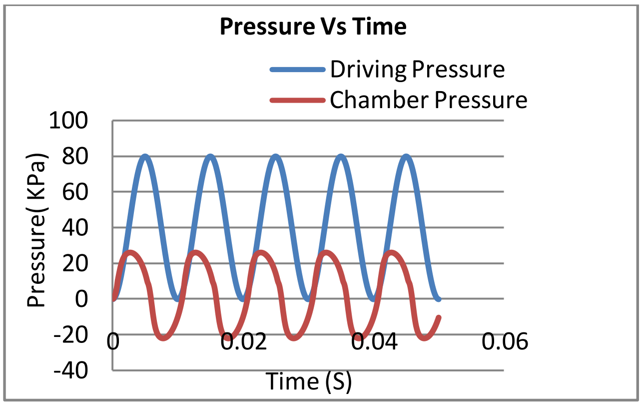

Fig 3: Microchamber_Pressure(Pch,Pdrive) Vs Time (graph)

B. *Stroke volume vs Time*

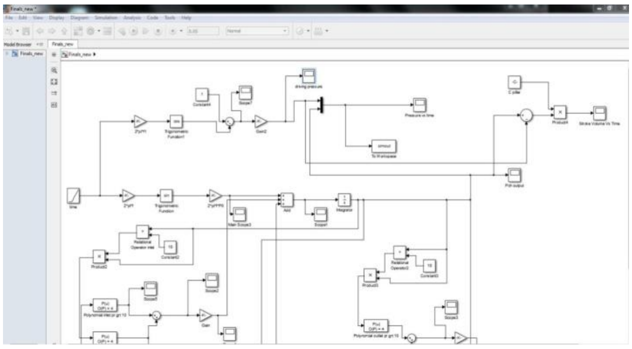

Fig 4 Microchamber Stroke volumeVs Time (graph)

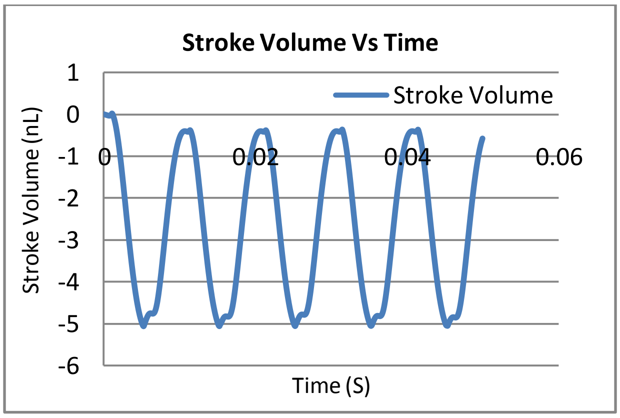

Fig 5 Microchamber_Stroke volume Vs Time (graph)

For getting the below figure , we need to simulate the equation Vch =Cpillar (Pch-P).So input for this equation is taken from respective lines if chamber pressure and driving pressure (Fig 4 and Fig 5)

C. *Flowrate Vs Time*

Since there is no flow loss in inlet valve,flow rate through outlet valve alone gives the net flow rate output. A scope connection to the outlet line of Φiv gives the flowratevs time graph (Fig 6 and fig 7).The negative side of the graph which shows the the back flow rate.

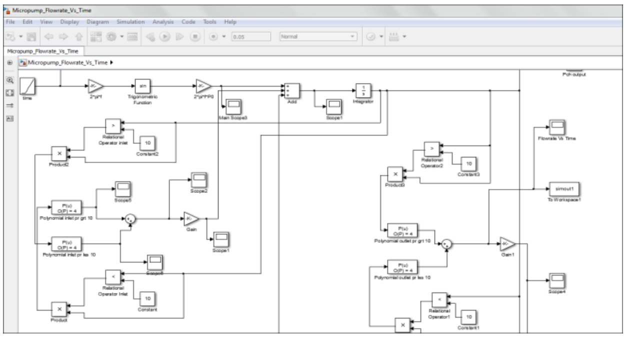

Fig 6 Microchamber_FlowrateVs Time

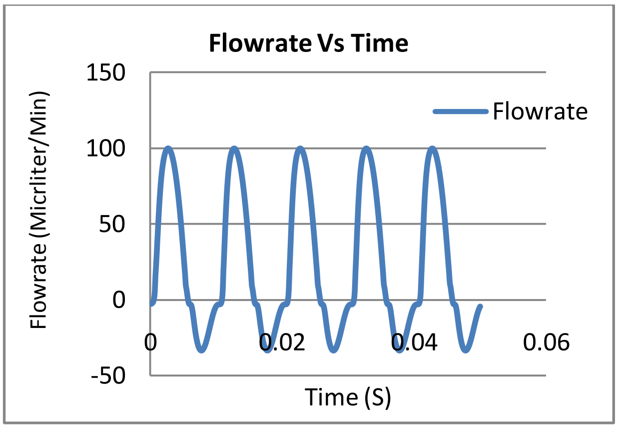

Fig 7 Microchamber_ FlowrateVs Time (graph)

D. *Flow Rate Vs Pressure Difference*

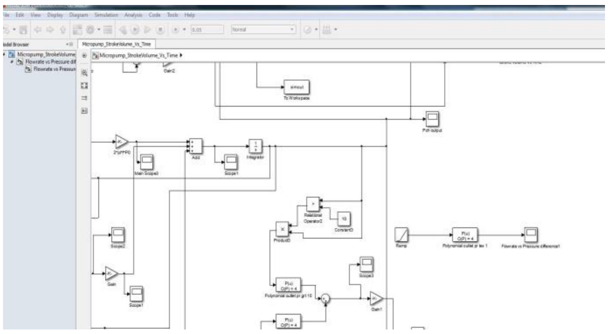

Fig 8 Microchamber Flow rate vs Pressure Difference

This graphical representation obtained by provide separate input signal to outlet valve polynomial(fig 7 and fig 8).This operation in Mat lab only shows the way through get the result is correct. Instead of providing the pressure difference in X axis, here we provide a Ramp signal as input having initial value is -7 and slope 240.This is obtained by the equation y=mx +c, since the variation in x axis is linear.

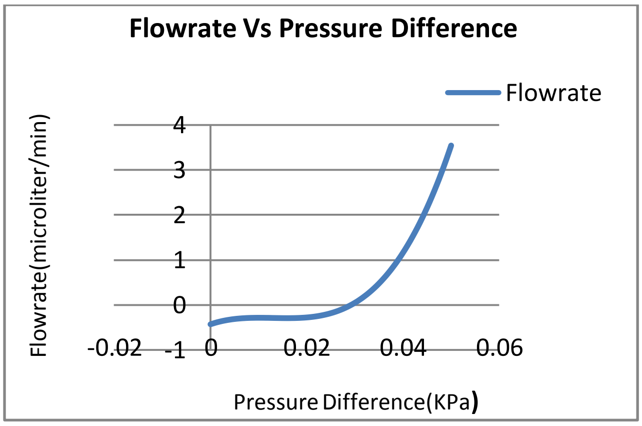

Fig 9 MicrochamberFlow rate vs Pressure Difference (graph)

V. CONCLUSION

The quasistatic ODE was finally solved by the calculated stroke volume of the membrane, the flow rates of inlet and outlet microvalves, and an assigned driving pressure By using the Mat Lab the governing equations from physical model is numerically simulated. In results analysis some of the variation is occur at pressure 10kpa, it is due to the fact that the polynomial variation is provided at the above mentioned pressure in inlet and outlet valve. When we are changing the constant provided at the relational operator that will affect on the area that mentioned earlier. The result thus obtained is almost correct since there is a negative flow rate due to the back pressure. Flow rate decreases with increase in back pressure. Chamber pressure is less than the driving pressure that provides and also chamber Pressure is in the form of sinusoidal wave. The outlet valve of microchamber become closed when the pressure in the microchamber become negative.

VI. ACKNOWLEDGMENT

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