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BEHAVIORAL STUDY OF A MICRO SCALES VALVELESS IMPEDANCE PUMP

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Abstract*- The implementation of Micro fluidic devices is difficult because when fluid flows through a small restricted passage it is intensely dominated by viscous dissipation. Fluid transport requires the ability to both pump and mix especially where transport must occur over long lengths within short time interval. The pump is therefore an integral component of any micro fluidic system. A valve less mechanism consists of an impedance pump which is a device that uses mismatch in acoustic impedance to drive flow. A behavioral study of such a pump is performed here by varying the parameter elastic tube length. With the designed actuator a length of 113mm of elastic tube is considered as the most feasible which gives the maximum stable flow rate.*

Keywords: Micro-fluidic devices, impedance, mismatch, elastic tube length.

1. INTRODUCTION

Within any micro environment a number of processes can be done by microfluidic system. The surface area to volume ratio of micro-fluidic systems offers an effective means for consumption and increased transport capabilities. In biological systems, the ability to transport fluid within the veins and arteries which is obviously a micro environment; simultaneously delivers fluids through different organs and nourishes the system by removing wastes. When fluid flows through a restricted area viscous force becomes dominant over the momentum force. Therefore transportation through this kind of passage requires thorough mixing and pumping. Hence the pump is an incomparable part of the microfluidic system. The benevolent characteristics of impedance pump make it an effective driving mechanism for micro fluidic systems. It uses a bio-inspired mechanism for valve less pumping based on resonant wave interactions along a flexible media the therefore can easily be manufactured. In comparison to the traditional electric pumps it requires no blades, rotors, high electric or magnetic fields which make it comfortable to bio-molecules. Flow direction can be reversed by changing the frequency or location of actuation. All of these characteristics make the impedance pump more than suitable for medical uses, for making suspensions, for mixing and pumping chemicals and moreover when micro in size for biomedical uses.

The valveless mechanism is interesting because it can be used to explain the complicated valveless blood flow mechanism in the circulation system of human body. Liebau demonstrated the various experiments of valveless pumping [1, 2, 3, 4] and observed a net flow from these experiments. His interest was motivated by the observation that sufficient blood circulation is retained in patients with probably inoperative aortic valves [5] and by the theory (William Harvey's concept of the circulation) that the heart alone is not good enough to obtain the necessary pumping, but it is assisted [6].

Besides an explanation of the valveless blood flow mechanism in the circulation, valveless pumping can also be found in the biomedical engineering, biology and engineering fields. For example, in early stages of the human embryonic circulation: At the end of the third week of the human fetus heart valves are not yet developed and the heart is beating in a coordinated fashion causing blood to circulate in a unidirectional way in the absence of heart valves [7,8].

1.1 Pump Background

Although impedance based pumping has been known about for some time [1-5] the dynamics are complex and the full potential has not been realized due to the wide number of parameters which can affect performance. The potential of impedance pumping as well as to demonstrate it is a robust and scalable concept which can be readily adapted for use biomedical devices and applications in microfluidics [9, 10] are illustrated in the PhD thesis of Derek Rinderknecht [11]. An extensive review of computational and experimental studies can be found in Chapter 1 section 2 of the PhD thesis of Anna Hickerson [12].

Several patents are registered using a valveless pump. The patents includes the use of a valveless pump in

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bypass grafts [13], to force blood through the graft without extending inside the graft, in an Intravascular diagnostic and Therapeutic sampling device [14], for uninterrupted blood sampling from a patient where the impedance pump is the driver in a micro-fluidic system and in stents [15], and in the Implantable Occular Pump to reduce intraocular pressure where the pump allows an adequate blood flow to prevent clotting within a lumen of the stent, and wherein the blood flow is equal to or greater than a blood flow at the implant site before implantation [16].

In this study a valveless impedance pump is designed and the behavior is investigated by varying the elastic tube length. The parameter elastic tube length was varied in the experiments of Anna Hickerson [9,10,12].The results are presented in flow rates and are compared to results found in other experimental studies, like those of Hickerson [9,10,12].

2. PRINCIPLE AND BASIC PHYSICS OF IMPEDANCE PUMP

The most fundamental principle of its operation is that the actuation occurs asymmetrically with respect to the impedance of the fluid system. Excitation and asymmetry are the two basic requirements for an impedance pump. Impedance pumps are usually formed by a channel or tube composed of a thin membrane of any material coupled on either side to wave reflection sites. When this thin membrane is coupled with either end to another material of different mechanical properties, geometries or any other factor which affects the wave propagation and/or reflection creates an impedance difference and therefore a site for wave reflection. Generally wave reflection site is created by the change of material property, geometry and asymmetry of length. Periodic pinching at an asymmetric location at a certain frequency, waveform and duty cycle results in the formation of a pressure gradient from wave interference and therefore the potential to drive flow. A schematic showing the impedance pump can be seen in fig. 2.1:

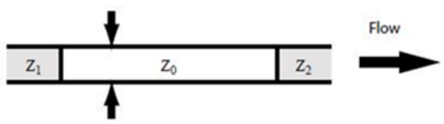

Fig. 2.1: A schematic of the impedance pump. [17]

The concept is described briefly as follows:
The area between Z_1 and Z_0 as well as Z_0 and Z_2 form the sites for wave reflection and the arrows represent the pinching location. By pinching at either side of the area of excitation provides the asymmetry of the system. Moreover pinching at different location of the length of excitation provides different wave property therefore different flow rate. The impedance pump is reversible, however before the reversal flow the first frequency range drive the flow from left to right side in the figure.

3. MATERIALS AND METHOD

Two different kinds of tubing are used for the elastic part and the rigid part. The elastic part consists of a latex tube of an inner diameter of 18 mm. The latex tube is connected to a T-junction which accommodates an injection site. The injection site is an IV drip, which are commonly used in a hospital. Water can be put into the setup and air can be taken from the setup at this injection site using a syringe. The other end elastic tube is connected to the PVC pipe of inner diameter 9.6 mm. A mechanical actuator is used in this experiment. It uses crank inspired mechanism. A small disc is inserted on the shaft of a DC gear motor. A plate is connected through nut and bolt joint with the disc. The other end of the rectangular plate is bolted with an end of a fulcrum arm which rests on a U shaped base. The base arm contains several holes for perpendicular adjustment through nut bolt joint. A bolt of 10 mm wide head is connected at the other end of the fulcrum arm. The arm contains several slots to adjust the pinch position horizontally. When the motor starts to rotate it puts the plate in up down motion which pushes the fulcrum arm up and down and as a result the bolt pinches the elastic tube and fluid inside the tube starts to flow. A flow sensor is attached to the PVC at the opposite direction of pump. The sensor is connected to an Arduino Uno board which is connected to the computer through USB port. From the Arduino software the output flow rate can be read. The experimental set up is shown in the following fig.

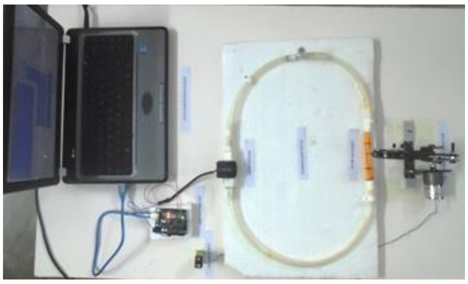

Fig. 3.1: The Experimental Setup

3.1 Experiments

Experiments are conducted to investigate the influence of elastic tube length on the pump's behavior. All experiments were performed at room temperature of approximately 298 K.

The volume inside the setup is kept constant for every experiment. First, the empty setup is weighted. After filling the setup with approximately the right volume of water, the setup is weighted again. The difference between the weight of the empty and full setup should then be equal to the desired volume, assuming that water has a density of 1000 kg/m^3, so that 1g of water corresponds to 1 ml of water. Before every experiment the weight of the setup is adjusted to the desired volume by adding/taking away water using a syringe. The parameter which is varied in this experiment is described below

3.1.1 Elastic Tube Length

The influence of the length of the elastic tube is examined at 9 different pinch locations between 26 and 74% from the left end of the elastic tube. The pinch frequency is equal to 5.25 Hz and the volume inside the

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setup is equal to 116 ml. The elastic tube length is adjusted by changing the elastic tube and PVC tube for several times.

3.2 Analysis of Results

The flow inside the setup is investigated in all experiments and is characterized using both dimensional and non-dimensional parameters. The dimensional parameters are described in the previous paragraph.

Characterizing the flow using non-dimensional parameters makes it possible to compare the results of these experiments to experiments done by others, like [9, 10, 12]. This is done in two different ways:

In the first, the incompressible Newtonian flow is characterized by the Womersley number, ά and the Reynolds number, Re

$$\alpha = R\sqrt{\frac{\omega}{\grave{\upsilon}}} \tag{3.1}$$

$$Re = \frac{Rv}{\grave{\upsilon}} \tag{3.2}$$

Where, R is the radius of the elastic tube, ὺ is the kinematic viscosity of the fluid inside the setup, ὼ is the angular frequency which corresponds to 2пf, where f is the pinch frequency that is regulated and registered by the mechanical pincher device, v is the fluid velocity inside the setup derived from the flow rate measured by the flow meter. This is done as follow:

$$v = \frac{Q}{A} \tag{3.3}$$

Where, A the surface of the cross-section of the elastic tube which is equal to πR^2, Q the flow rate in m^3/s measured by the flow meter.

The values of the parameters used in the above Eq. (3.1), (3.2) and (3.3) can be found in Table 3.1

Table 3.1: Overview of the parameters used to determine the characteristic numbers ά and Re

Parameter	Value
Inner Radius, R	5×10^{-3} m
Angular velocity, ὼ	7 rad/s
Viscosity of water, ὺ	0.00103 kg/m.s
Cross sectional area of the tube, A	78.54×10^{-6} m^2/s
Velocity inside tube, v	Follows from experiment

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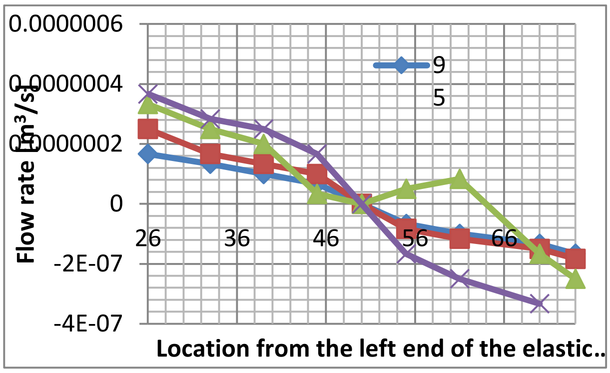
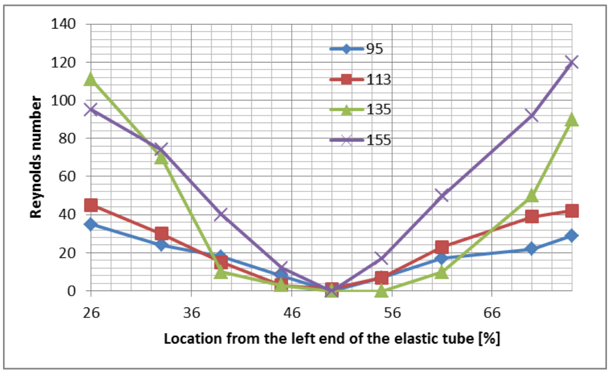
4. RESULTS

The results of the experiment can be seen in fig. 4.1. What can be seen in fig. 4.1(a) is that the magnitude of the flow rate decreases when the pinch location is closer to the center of the elastic tube. Flow reversal can be seen when the pinch location is chosen to the right side of the center of the elastic tube. Furthermore, the flow rate is close to zero at the center of the elastic tube for all tube lengths. Moreover, the magnitude of the flow rate decreases when the length of the elastic tube is decreased. However, the flow rate for the 135 mm tube length was higher in magnitude at the 26% pinch location and approximately equal to the flow rate of the 155 mm tube at the 32% pinch location. The 155 mm tube had the highest flow rate magnitudes for all the other pinch locations. Furthermore, the flow rate of the 135 mm tube was lower in magnitude than the two smaller tubes at pinch location 39, 45, 55 and 61%. The magnitude of the flow rate of the 113 mm tube is higher than the magnitude of the flow rate of the 95 mm tube, except at the 39% pinch location where it is the other way around. The same can be seen in fig. 4.1(b)



(a)



(b)

Fig. 4.1: Results for the elastic tube length experiment. Flow rate versus pinch location for elastic tube lengths 95, 113, 135 and 155 mm, both dimensional (a) and non-dimensional using Re (b)

5. CONCLUSION

A number of publications have been written trying to model the behavior of a valveless impedance pump. Nevertheless, except from Hickerson [9, 10, 12], little has been done experimentally to determine the nature of pumping being modeled. Presented in this study are the experimental results that provide insight into the dominant parameter that is crucial in the mechanism of the valveless impedance pump. Flow responses under the effect elastic tube length have been measured which is not done previously. It is evident from the study that the flow rate increases when the length of the elastic tube is increased.

6. REFERENCES

[1] G. Liebau, "Uber ein Ventilloses Pumpprinzip", *Naturwissenschaften*, vol. 41, pp. 327-328, 1954.
[2] G. Liebau, "Die Stromungsprinzipien des Herzens",

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Zeitschrift Kreislaufforschung, vol. 44, pp. 677-684, 1955.

[3] G. Liebau, "Aus welchem bleibt die Blutforderung durch das Herz bei valvularem Versagen erhalten?", *Zeitschrift Kreislaufforschung*, vol. 45, pp. 481-488, 1954.

[4] G. Liebau, "Die Bedeutung der Tragheitskrafte fur die Dynamik des Blutkreislaufs", *Zeitschrift Kreislaufforschung*, vol. 46, pp. 428-438, 1957.

[5] H. Thomann, "A Simple Pumping Mechanism in a Valveless Tube", *Journal of Applied Mathematics and Physics*, vol. 29, pp. 169-177, 1978.

[6] W. Harvey, *Exercitatio Anatomica de Motu Cordis et Sanguinis in Animalibus* , Chapter 14, Frankford 1987.

[7] K.L. Moore, *Embryologie:Lehrbuch und Atlas der Entwicklungsgeschichte des Mens*, pp. 340-358, Schattauer, Stuttgart. 1985.

[8] A.S. Forouhar, M. Liebling, and A. Hickerson, "The embryonic Vertebrate Heart Tube is a Dynamic Suction Pump", *Science*, vol. 312, pp. 751-753, 2006.

[9] A. I. Hickerson, D. Rinderknecht, and M. Gharib, "Experimental study of the behavior of a valveless impedance pump", *Experiments in Fluids*, vol. 38 no. 4, pp. 534-540, 2005.

[10] D. Rinderknecht, A. I. Hickerson, and M. Gharib, "A valveless micro impedance pump driven by electromagnetic actuation", *Journal of Micromechanics and Microengineering*, vol. 15, no. 4, pp. 861-866, 2005.

[11] D. Rinderknecht, *Development of a micro-impedance pump for pulsatile flow*, PhD thesis, California Institute of Technology 2008.

[12] A. I. Hickerson, *An experimental analysis of the characteristic behaviors of an impedance pump*, PhD thesis, California Institute of Technology, 2005.

[13] M. Gharib, D. Rinderknecht, and I. Avrahami, "*Impedance pump used in bypass grafts*" 2007 http://www.freepatentsonline.com/20070038016.html (accessed 15 January, 2015)

[14] H. Tu, D. Haffner, and M. Gharib, "*Intravascular diagnostic and therapeutic sampling device*". 2005 http://www.freepatentsonline.com/20050049578.html (accessed15 January, 2015)

[15] M. Gharib, D. Rinderknecht, and D. Petrasek, "*Intravascular miniature stent pump*", 2006 http://www.freepatentsonline.com/20060280655.html (15 January, 2015)

[16] M. Gharib, A. Iwaniec, R. A. Wolf, "*Implantable occular pump to reduce intraocular pressure*", 2003 http://www.freepatentsonline.com/20030233143.html (accessed 15 January, 2015)

8. NOMENCLATURE

Symbol	**Meaning**	**Unit**
R	Radius of the elastic tube	(m)
ὼ	Angular velocity	(rad/s)
A	Cross sectional area of the tube	(m^2)
ὺ	Kinematic viscosity	(kg/m.s)
v	Velocity inside tube	(m/s)
Q	Flow rate	(m^3/s)
ά	Womersley number	Dimensio-nless
Re	Reynolds number	Dimensio-nless

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