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MEASUREMENT OF FLOW IN CIRCULAR ELASTIC TUBE

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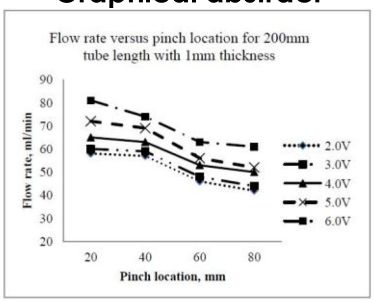
Graphical abstract



Abstract

Impedance pump is a simple valveless pumping mechanism which typically used in viscosity measurement device to assist pumping of fluid. It is typically connected to an elastic tube in a circulatory system of a more rigid tube. In conventional mechanical circulatory support systems using rotary pump, the pumping mechanism was exposed to turbulent stresses. Hence, this may cause damage to blood cells flowing through the impeller. There has been initial work on finding alternative solution using the impedance pump system. However, substantial findings are not yet sufficient to fully understand the mechanism. The purpose of this research is to extend the investigation on impedance pump by specifically looking at the effect of structural parameters on the elastic tube and the flow behaviour. In this study, a closed loop impedance pump system was set up to demonstrate blood flow circulatory system where the mixture of glycerine and water was used as the working fluid. Three variables were regulated namely voltage, tube thickness, and tube length was used in order to get the flowrate of the working fluid. Based on the results, it was found that tube thickness of 1 mm and a length of 200 mm had produced the highest flowrate in the region 75 ml/min.

Keywords: Valveless pump, lastic tube, impedance pump, thickness, flow rate

1.0 INTRODUCTION

The blood circulatory system in human body consists of blood as the working fluid, blood vessels, and heart. The main functions of the circulatory system are to transport nutrients to the outlying parts and assist in inhalation of oxygen and exhalation of the waste products like carbon dioxide via lungs [1]. Once the circulation system fails, it causes the rest of the organs in the body to malfunction. In this situation, the primary part that is adversely affected is the heart. This will lead to chronic heart disease and eventually heart failure. As a result, the heart is unable to pump sufficient blood throughout the body due to insufficient pumping force [2]. A typical remedy for a patient having this heart failure is to have a heart transplant. These patients often use a heart assist device as temporary measures while waiting for compatible donors. However, many patients died and suffered while waiting for heart transplant due to lack of donors [3]. In America, the United States Food and Drug Administration has approved New Generation Heart Mate II as a temporary supporting device before heart donation and transplant process takes place as well as for destination therapy or long-term [4]. The device is controlled by axially rotating disc pump supported with a valve mechanism. The main concern of the mechanism is that it tends to produce viscous and turbulent stresses. It may also cause damage of blood cells flowing through the impeller.

Liebau suggested that viscosity, inertia and elasticity tend to affect the parameters measured in the device [5]. One of the potential approaches to overcome the situation is to use mechanical device that can assist blood flow in patient suffering from heart problem such as the loss of blood pressure and low cardiac output. An existing device used to help heart failure patients is called Ventricular Assist Device VAD. Multiple different mechanical devices for long-term circulatory support have been developed, ranging from total artificial hearts to ventricular assist devices (VADs). The main purpose of a VAD is to unload the failing heart and help maintain forward cardiac output and vital organ perfusion. Originally VAD was introduced as a temporary bridge during recovery and then as a bridge for heart transplant. Over time, VADs have involved into a permanent or "destination" therapy for a growing number of patients with refractory heart failure [6]. VADs are mechanical circulatory devices deployed to partially or completely replace the function of failing hearts. Some VADs are intended for short term use, typically for patients recovering from heart attacks or heart surgery, while others are intended for long-term use (months to years and in some cases for life), typically for patients suffering from congestive heart failure [7]. With the advance in technology, the possibility of surviving the heart attact among the patients has also increased. However the current VODs still have some limitation.

The current generation of mechanical pumps requires thinning of the blood (or anticoagulation) to prevent the formation of blood clots. These clots can cause dysfunction of the pump itself or more likely can embolize into the circulation and cause a transient ischemic attack, stroke, or lack of blood flow to a limb or vital organ [8]. This pumping concept offers a low energy, low noise alternatively at both micro and macro scales [9]. Valveless pumps are easy to construct and require few moving parts [10]. There are lots of characteristic of impedance pump that can be discovered. By obtaining the flow rate for each different case, the best parameters which may provide the optimum flow rate can be identified. There has been an initial work on finding alternative solution using the impedance pump system. However, substantial findings are not yet sufficient to fully understand the mechanism. The purpose of this research is to extend the study on impedance pump by specifically looking at the effect of structural parameters on the elastic tube and the flow behaviour.

2.0 Methodology

The layout of the experimental set up is deployed in the current work as shown in Figure 1. The main components of the testrig are DC Motor, tee barb, actuator, flow meter, elastic tube and rigid tube. Initially, an elastic tube of 200mm long and 1mm thickness was connected into the circuit. The water pipe was completely filled into the whole circuit via T-tubing barb. The pincher location was first set at about 20 mm from the left side of the elastic tube and the voltage of DC motor was 2.0V. When the pincher pressed the elastic tube, the water inside the rigid tube flowed through the flow meter and the flow rates were recorded. The mesurement of flowrate was conducted for varying voltage values (3.0, 4.0, 5.0 and 6.0V). The test was repeated for different specifications as shown in Table 1.

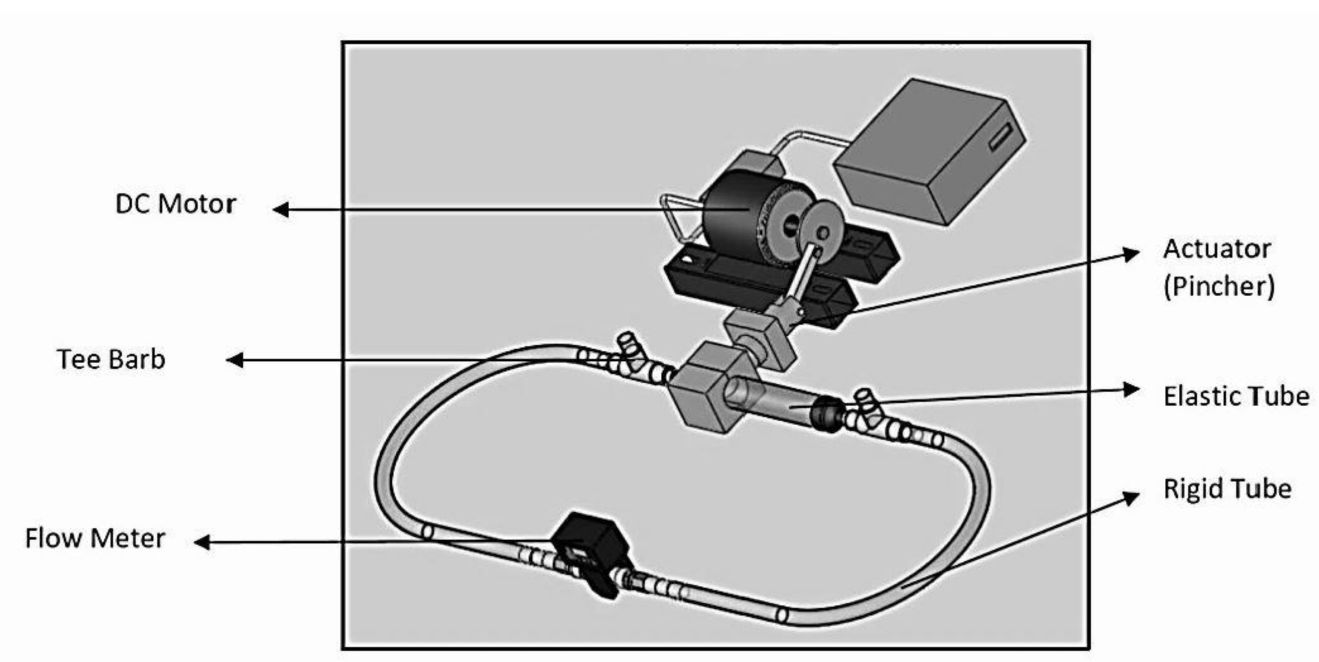


Figure 1 Experiment setup

Table 1 Detail specification of experiment variables

Part	Parameter	
Elastic tube	Outer Diameter, mm	30.0
	Thickness	1.0 and 3.0
	Length, mm	200.0 and 260.0
Actuator	Length, mm	150.0
	Diameter, mm	10.0
	Thickness, mm	10.0
Electrical DC motor	Voltage, V	2.0, 3.0, 4.0, 5.0 and 6.0

3.0 RESULT AND DISCUSSION

Figure 2 shows the dependence of flow rates on different thicknesses of elastic tubes with different voltage values. They were two samples of different thickness with the same length of 200 mm tested.

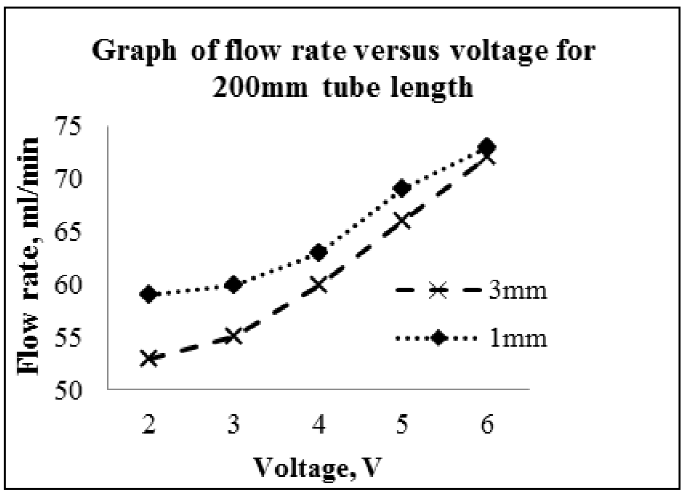


Figure 2 Effect of voltage on flow rate for 200 mm long tube

In Figure 2, the profile of flow rate over voltage is similar for both samples. It shows that flow rate increases with voltage. These voltage values represent pinching frequency. Theoritically, the higher the pinching frequency, the faster the flow rate. Base on Figure 2, the profile obtained follows the theoretical prediction. The profile suggests that at higher pinching frequency, the change in the flow rate is high as evident from the slope of the profile. However, it was found that the limiting voltage was 6 V beyond which the measurement of flow rate was no longer valid. This is because by increasing the voltage, it causes the rotational speed of the motor to increase.

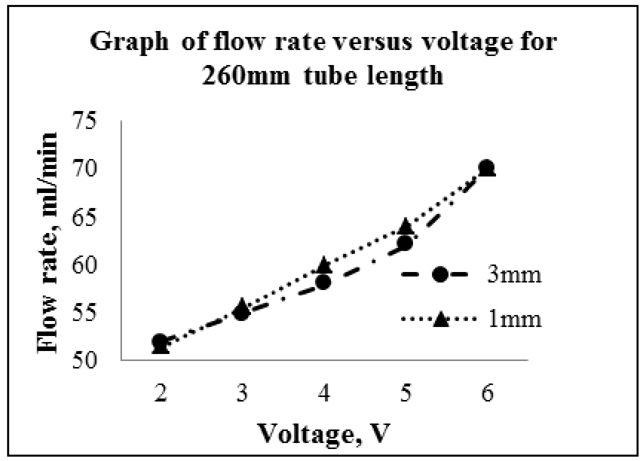


Figure 3 Effect of voltage on flow rate for 260 mm long tube

Hence, the kinetic energy delivered towards the elastic tube automatically increases forcing the working fluid to flow faster. The difference in the flow rate values for 1 mm and 3 mm thickness become smaller as voltage increases. This suggests that the effect of tube thickness become less at higher pinching frequency.

On the effect of thickness, the flow rate values observed for thicker tube were lower than those for thinner tube. This is because, as the thickness increased, the inner diameter of the tube decreased propotionately. The decrease in the inner diameter has affected the area of the elastic tube. The area of the specimen is directly proportional to the volume flow rate of the working fluid. Therefore, as the area decreased, the volume flow rate of the fluid also decreased.

The profiles of flow rate over voltage for 260 mm long tube are shown in Figure 3. It was observed that varying the thickness (1 mm and 3 mm) of the tube has less significant influence on the flow rate for the case of 260 mm long tube. This was different from what has been observed for 200 mm long tube. At longer tube length (260 mm), the non linear characteristics of the profile remain.

The correlation between flow rate and voltage (pinching frequency) for four different pinch locations is shown in Figure 4. This was done for 1 mm thick and 200 mm long tube because of the higher flow rate it produced as shown in Figure 2 and 3. In all cases of pinching locations, the profiles are similar.

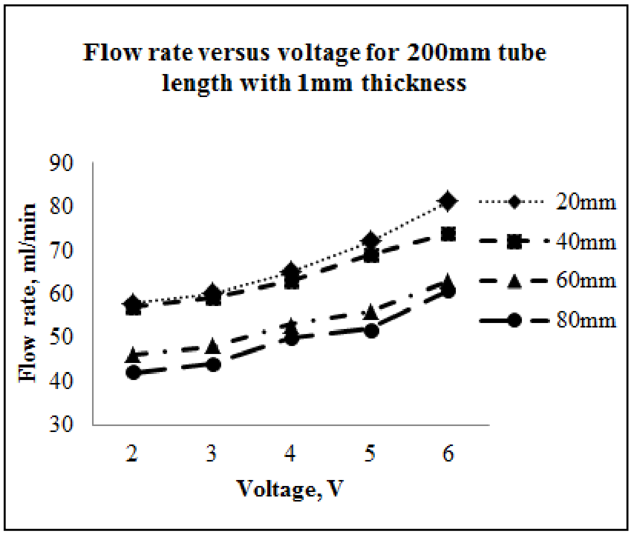


Figure 4 The effect of pinch locations on flowrate

In Figure 4, when pinch location is set near the middle of the elastic tube, flow rate measured around 40 ml/min for 2 V. As the voltage increases to 6 V, at the same location of the pinch, fluid flow rate is about 55 ml/min. Base on observation on the effect of pinch location, the nearer pinch location to the center of the elastic tube, the less flow rate are being recorded.

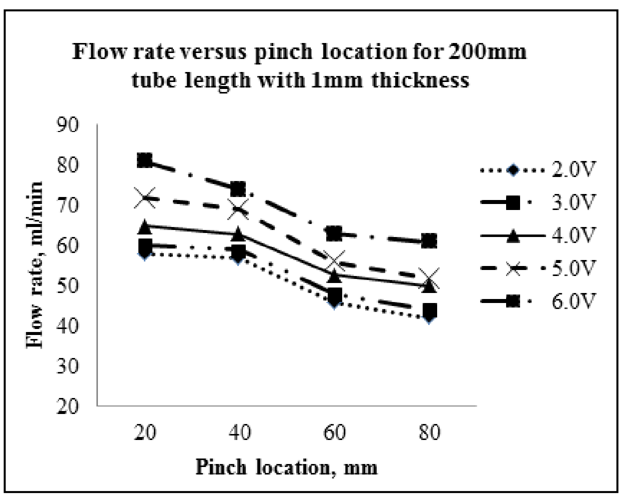


Figure 5 Effect of pinch locations

Figure 5 shows the dependence of flow rate on different voltage with varied pinch location. The elastic tube used for this test is 200 mm length and 1mm thickness respectively. The profiles of flowrate over voltage in Figure 5 have similar trend for all casses of pinching locations. It was found that pinch location has affected the flowrate to some extent. The closer the pinch location to the center of the elastic tube, the higher the flowrate. The lowest flow rate recorded at 6 V maximum setting was 60 ml/min. It was consistently found that higher voltage(pinching frequency) setting results higher fluid flow rate. Hence, the flow rate can be controlled by adjusting voltage supply or the pinch location of elastic tube that directs the motion of the fluid.

4.0 CONCLUSION

This research shows the impact of varying the thickness of the wall and length of the elastic tube on volume flow rate of the working fluid. Based on the results obtained, it was found that flowrate decreases with thickness and length of elastic tube. The elastic tube with thickness of 1mm and length of 200mm and the pinch location at 20mm resulted the highest flow rate of 81ml/min at applied voltage of 6.0V. In the experimental set up, it was found that voltage could only go up to 6 V beyond which the flow started to reverse giving a negative flowrate value. Another significant observation is that there was no flow rate recorded when the elastic tube was pinched at the middle due to the equivalent pressure values on both sides of the tube.

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