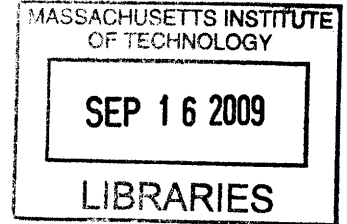# Drag Reducing Polymers as Simple Indicators of Hemolytic Potential in Biomechanical Devices

by

Sarah Shieh

Submitted to the Department of Mechanical Engineering
in Partial Fulfillment of the Requirements for the Degree of
Bachelor of Science in Mechanical Engineering

at the

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Chairman, Undergraduate Thesis Committee

1

Drag Reducing Polymers as Simple Indicators of Hemolytic Potential in Biomechanical Devices

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ABSTRACT

An experimental study was carried out to determine if drag reducing polymers can be simple indicators of hemolytic potential in biomechanical devices. Specifically, three different blood pumps, known as a left ventricle assist devices (LVADs) were operated in a test loop using an aqueous solution of polyethylene glycol (PEO, MW = 5000 kDa), a known drag reducing polymer. The pumps were operated under controlled parameters and the change in viscosity (cP) and drag reduction (%DR) for each pump was monitored over the specified time period. The CentriMag® (CM) was used to confirm the drag reducing behavior of PEO, while HeartMate® II (HM II) and HeartMate® III (HM III) were used to determine if there was a correlation between experimental results and actual hemolysis results.

Experimental results showed that the mathematical difference between the average final and initial viscosity of HM II was greater than the difference for HM III. HM II had a difference of 0.21 cP and HM III had a difference of 0.16 cP. Hemolysis results using bovine blood showed that HM II had a higher hemolysis rate of 3.80 ± 1.11 g/day and a higher milligram normalized index of hemolysis of 0.0393 ± 0.0155. The average hemolysis rate for HM III was 1.38 ± 0.63 g/day and the milligram normalized index of hemolysis (mg N.I.H.) was 0.571 ± 0.333. This positive correlation shows that PEO can be a simple indicator of hemolytic potential for biomechanical devices. More data and experimentation is needed to further understand the behavior of PEO and it's ability to indicate hemolytic potential using a wider range of biomechanical devices.

Thesis Supervisor: Anette E. Hosoi

Title: Professor of Mechanical Engineering

BIOGRAPHICAL NOTE

Sarah Shieh is currently a senior at the Massachusetts Institute of Technology. She is working towards her Bachelor of Science in Mechanical Engineering and plans to complete her four years of undergraduate work in June 2009. She plans on attending the Georgia Institute of Technology for her Master of Science in Mechanical Engineering, with a concentration in Bioengineering, starting in August of 2009. Sarah has a strong interest in designing medical devices, in hopes of bringing better biomedical technology to people.

ACKNOWLEDGEMENTS

1. INTRODUCTION

Hemolysis is one of many important considerations during the design of mechanical blood pumps intended to assist or replace a failing heart's pumping function. To assess the hemolytic potential of a particular blood pump design, testing is typically conducted with a variety of methods that offer increasing levels of effort and risk with increasing correlation to an actual human response. These methods include: computational fluid dynamic analysis of shear forces, thermal analysis, acquiring animal or human blood for *in vitro* testing, implanting a pump into an animal for *in vivo* testing or ultimately implanting a pump into a human during a clinical trial. The more accurate and challenging hemolysis testing will often be deferred until a design is mature, possibly influencing a design away from novel ideas or delaying the discovery of a suboptimal design.

Drag reducing polymers (DRPs) have been used in many applications where minimizing frictional losses in fluid transfer are desired, such as pumping oil through pipelines. An aqueous solution of these large molecules is actually quite viscous, however, when the solution is subjected to shear and/or heat the molecules break down and the solution becomes less viscous. This mechanism, though not desirable in some applications, may be beneficial in the earlier stages of designing biomechanical devices as a simple comparative indicator of hemolytic potential.

Currently, there is little information about using DRPs in this context. For that reason, this study is an attempt to determine the cross-correlation between drag reducing polymer degradation and blood pump hemolytic potential due to applied shear forces and/or heat. Polyethylene oxide (PEO), a polymer suggested in a previous study [1], is chosen as a potentially appropriate DRP. The previous study results will be verified using the CentriMag, by repeating all indicated conditions and pump settings. Then simple aqueous solution containing the DRP will be placed into two differently designed left ventricle assist devices, both of which are designed by Thoratec Corporation (Pleasanton, CA). Previous hemolysis results for the two pumps using bovine blood will be compared to the experimental results using the DRP. Ultimately, the hope is to determine the ability of drag reducing polymers to assist the design of biomechanical devices by serving as a simple indicator of hemolytic potential that can be easily employed at any stage of the design process.

2. BACKGROUND

2.1. *Fluid Flow*

There are two main fluid flow regimes: laminar and turbulent. Laminar flow occurs when a fluid flows in parallel lines, with no disruption between the layers. This flow regime is characterized by high momentum diffusion, low momentum convection, pressure and velocity independent from time. The shape of the velocity profile across at the section of a pipe for laminar flow is a parabolic shape, with the maximum velocity at the center being about twice the average velocity of the pipe. Opposite to laminar flow is turbulent flow. In the turbulent regime, a fairly flat velocity distribution exists across the section of the pipe, as shown in Figure 1. As a result, the entire fluid flow can be described by a given single value. The velocity of the fluid in contact with the pipe is essentially zero and increases with increasing distance from the wall.
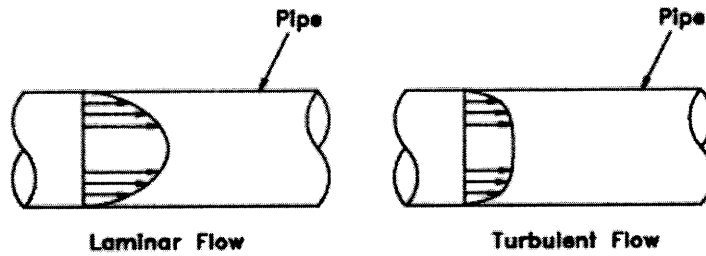
Figure 1. Velocity profiles for laminar and turbulent flow. Laminar flow has a parabolic velocity profile, with the maximum velocity located at the center of the pipeline. The turbulent flow has a fairly flat velocity distribution across the section of the pipe.

When a fluid comes into contact with a solid surface, such as a pipe wall, it experiences a resistance, known as drag or frictional pressure drop. The frictional forces that resist the flow of the fluid induce a large pressure drop. This pressure drop is greatest when there is turbulent flow, which has been shown in a study conducted by Sung J. H., et al [2]. Therefore, in order to maintain the same volumetric flow undergoing turbulent flow in a pipe, a low pressure drop is necessary.

2.2. *Toms Effect*

The increase in drag has serious implications in design of piping systems where the energy required for pumping turbulent fluids increases considerably relative to laminar flow. In 1948, Toms discovered that the addition of minute quantities of high molecular weight polymers to a Newtonian fluid in turbulent flow could considerably reduce frictional drag exerted by the fluid when it flows over a surface, such as a pipe [3]. Following Toms discovery, Kulicke was able to confirm Toms discovery as well as show that the significant decrease in resistance to flow did not greatly affect the viscosity or density of the flow [4].

2.3. *Dynamic Viscosity*

Dynamic viscosity is the measure of the resistance of a fluid, which is being deformed by either shear stress or axial stress. It is the tangential force per unit area required to move one horizontal plane with respect to the other, at unit velocity, while maintaining the planes at a unit distance apart. Figure 2 is a diagram that illustrates the meaning of dynamic viscosity.

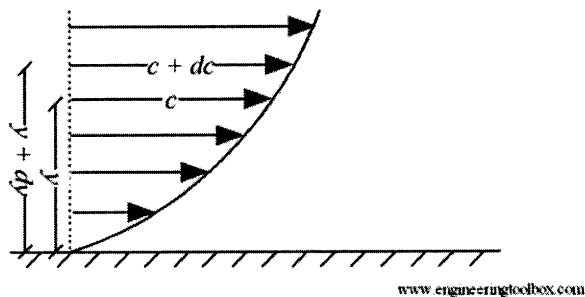
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Figure 2: Dynamic viscosity. It is the tangential force per unit area required to move one horizontal plane with respect to the other, at unit velocity, while maintaining the planes at a unit distance (dy) apart.

The dynamic or absolute viscosity can be expressed as:

$$\tau = \mu \frac{dc}{dy} \qquad ,$$ (1)

where τ is the shearing stress and μ is the dynamic viscosity.

Viscosity is always not a constant value for a fluid. One factor that affects viscosity is temperature. Typically, the viscosity of the fluid decreases with increasing temperature. However, the sensitivity to temperature varies from fluid to fluid. At the molecular level, as temperature increases, the average speed of the molecules in a liquid also increases. In turn, the amount of time the molecules are in contact with each other decreases. Therefore, as temperature increases, the average intermolecular forces decrease, resulting a lower viscosity at higher temperatures.

2.4. *Characteristics of DRPs*

The origin of the drag reduction mechanism is the stretching of polymer molecules in the turbulent flow. This stretching dampens turbulent fluctuations, thereby reducing drag and the amount of energy expended [5]. The effectiveness of the polymer stretching is a function of the size of the polymer, which is governed by its molar mass, and the number of polymer molecules, which is governed by the polymer concentration.

It has been shown that the most effective DRPs are those that have a large molecular weight. The average molecular weight is typically greater than one million Daltons. As the molecular weight of the polymer increases, the ability of the polymer to reduce drag also increases. Additionally, these polymers typically have a long flexible backbone, a high degree of polymerization, and a fairly linear structure [6]. Examples of such polymers include poly(ethylene oxide) (PEO), poly(acrylamide) (PAAM), and polyisobutylene (PIB) [7].

2.5. *Applications*

The study of DRPs is important because it has several applications in engineering. The first application of DRPs was in piping systems. When fluid flows through pipelines, such as during the transferring of oil, the flow is typically turbulent. This means that as the fluid is pumped through the pipes, frictional losses increases. As the frictional losses increases, more energy is required to pump the fluid, which is inefficient and costly. However, since the discovery of DRPs, these polymers have been placed into these pipes as a method to reduce the pumping energy. In some cases, injecting a low concentration of about 5-10 ppm stock solution of the polymer can promote a drag reduction of 50%. [8]

The use of DRPs has also been applied to biomedicine. Although turbulent conditions are not present in vascular systems, when nanomolar concentrations of blood soluble DRPs were injected intravenously, there was an increase in aortic and arterial blood flow and a decrease in both blood pressure and peripheral vascular resistance, with no direct effect on blood viscosity or blood vessel tone [9, 10]. Additionally, when the polymer was inserted intravenously, there was a significant increase in collateral blood flow in rabbits [11] and an increase in the number of capillaries in normal and diabetic rats [12].

Another biomedical application of these polymers is that they have been used to diminish mechanical damage to blood cells when the blood is in contact with circulatory assist devices. A previous study showed that polyethylene glycol (PEG) could be a potential erythrocyte protective agent against mechanically inducted cellular trauma. A PEG concentration as low as 0.1% reduced hemolysis by more than 40% [13].

2.6.   *Limitations*

While the applications of DRPs are very broad, there are also limits to the performance of these polymers. The extent of drag reduction and shear stability are the two important factors that decide the efficacy of the polymer used for reducing the frictional loses under turbulent flow conditions. The extent of drag reduction increases as the molecular weight of the polymeric additive increases. Poor shear stability has been a major drawback for the commercial application of the phenomenon of drag reduction [14]. As the large molecules experience shear stresses, they degrade, thereby losing their ability to act as drag reducers. Shear stresses can occur in all systems, including flow through pipes or biomechanical devices.

2.7.   *Hemolysis*

Hemolysis is the destruction of red blood cells (RBCs) and the subsequent release of hemoglobin into the blood stream. As shown by a previous study, the destruction results from an interaction of mechanical forces: shear rate and pressure [15]. These damaging effects are dependent on shear forces and exposure times. Since such mechanical forces are present in biomechanical devices, such as heart pumps, designing such devices to not induce hemolysis is very important.

According to the ASTM standard of assessing hemolysis (Standard Practice for Assessment of Hemolysis in Continuous Flow Blood Pumps, F1841-97), the normalized index of hemolysis (N.I.H.) is the added grams of plasma free hemoglobin per 100 l of blood pumped, corrected for plasma volume using hematocrit and normalized by flow rate and circulation time. [16] Normalized milligram index of hemolysis (mg N.I.H.) the N.I.H. expressed by milligram value of free plasma hemoglobin. The equation for mg. N.I.H. can be found in Equation 2.

$$mg\,N.I.H.(mg/100l) = \Delta freeHb \times V \times \frac{100-Ht}{100} \times \frac{100}{Q \times T} \quad , \quad (2)$$

where $\Delta freeHl$ is the increase of plasma free hemoglobin concentration (g/L) over the sampling time interval. Additionally, V is the circuit volume (L), Q is the flow rate (L/min), Ht is the hematocrit (%), and T is the sample time interval (min). When in comparison, a higher value in mg N.I.H. indicates more hemolysis.

2.8. *Left Ventricle Assist Device (LVAD)*

The left ventricle is a large, muscular chamber of the heart. It pumps blood from the left ventricle into the aorta, which then distributes the blood to the rest of the body. LVADs can either be used as a temporary replacement for patients waiting for a heart transplant or for permanent use. In either situation, the already-weakened heart cannot

take any additional stress. The surgically implanted pump relieves the patient's heart and performs the pumping.

A typical type of LVAD has a tube that draws blood from the left ventricle into a pump. The pump then sends blood into the aorta, which is the large artery leaving the left ventricle, thereby restoring perfusion to the body. The pump is placed in the upper part of the abdomen. Another tube attached to the pump is brought out of the abdominal wall to the outside of the body and attached to the pump's battery and control system. The flow path of blood and the placement of the LVAD can be seen in Figure 3.


a. Inflow conduit
b. Outflow graft
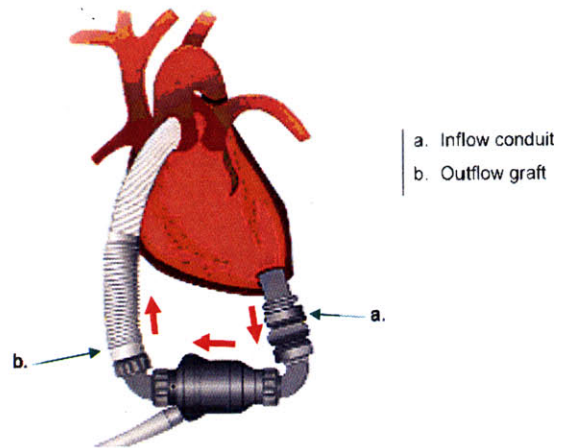
HMII Operating Manual [17]

Figure 3: Blood flow in and placement of a LVAD. A LVAD takes blood from the left ventricle and pumps it into the aorta. The device is placed in the upper part of the abdomen, where it is connected to the left ventricle.

Patients with LVADs can be discharged from the hospital and have an acceptable quality of life. According to an abstract presented at the American Heart Association's 2005 Scientific Sessions, LVADs reduced the risk of death in end-stage heart failure patients by 50 percent at six and 12 months and extended the average life span from 3.1 months to more than 10 months [18]. Recent discoveries have even shown that using a LVAD may actually help restore the heart into a normal healthy state. In a study published in Circulation in 2005, LVADs restored failing hearts in some patients with heart failure, allowing the pump to be removed and eliminating the need for a transplant.

2.8.1. HeartMate® II

The HM II is a product of Thoratec Corporation (Pleasanton, CA). It is intended for patients with end stage heart failure. The HM II LVAD is an axial-flow rotary ventricular assist device composed of a blood pump, percutaneous lead, external power source, and system driver. As shown in Figure 4a & 4b, it is attached between the apex of the left ventricle and the ascending aorta and is designed for long-term use. The system can be configured for battery power or tethered to an AC-supplied power base and monitor unit (Figure 5a & 5b, respectively) [19].
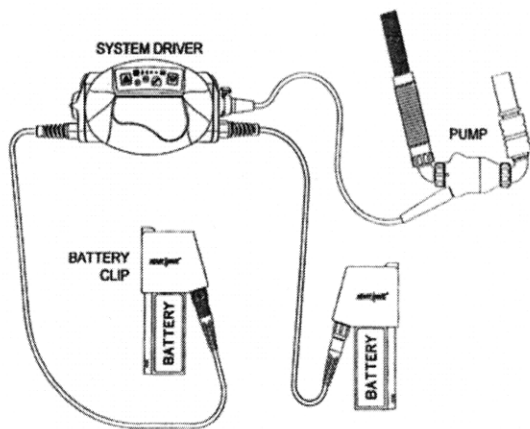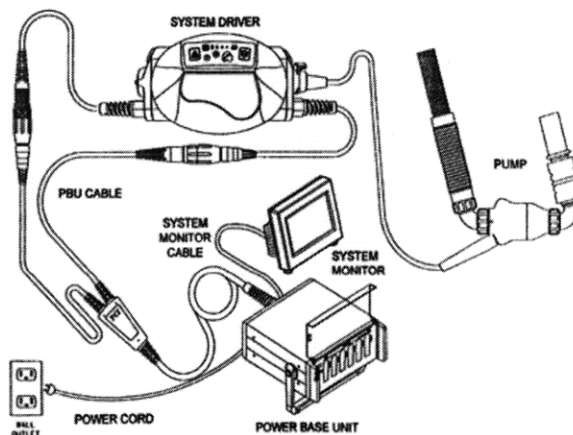
4a.
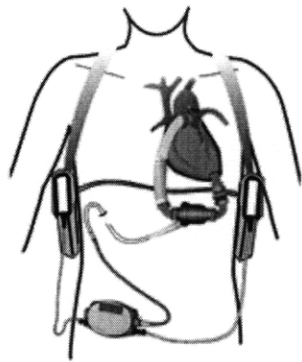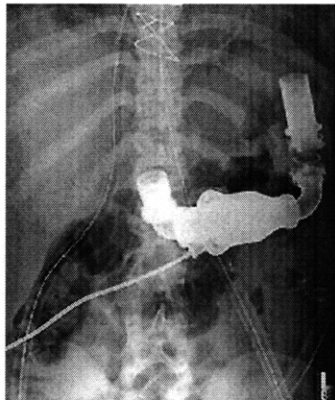
http://www.heartfailure.org/eng_site/whats_ne
w_HeartMate_II.asp

4b. [17]

Figure 4a and 4b. Placement of HM II. 3a (left) shows the placement of the HMII, along with its power cord, percutaneous cable, and batteries. 3b (right) is an x-ray picture, showing the attachment of HMII to the apex of the left ventricle and the ascending aorta and is designed for long-term use.

5a.

5b.

Figure 5a & 5b [17]. Battery versus AC power cord and monitor unit. 5a (left): HM II attached to two batteries. 5b (right): HM II attached to a power cord and monitor unit.

The HM II is a continuous-flow, axial-type rotary blood pump. Due to its small size and light weight, it is suitable for a wider range of patients, including children and small adults. The pump has a screw type rotor that contains a magnet, which can be seen in Figure 6. The rotor assembly is actuated by the electromotive force generated by the motor. The device operates at a constant speed which can be adjusted though a monitor.
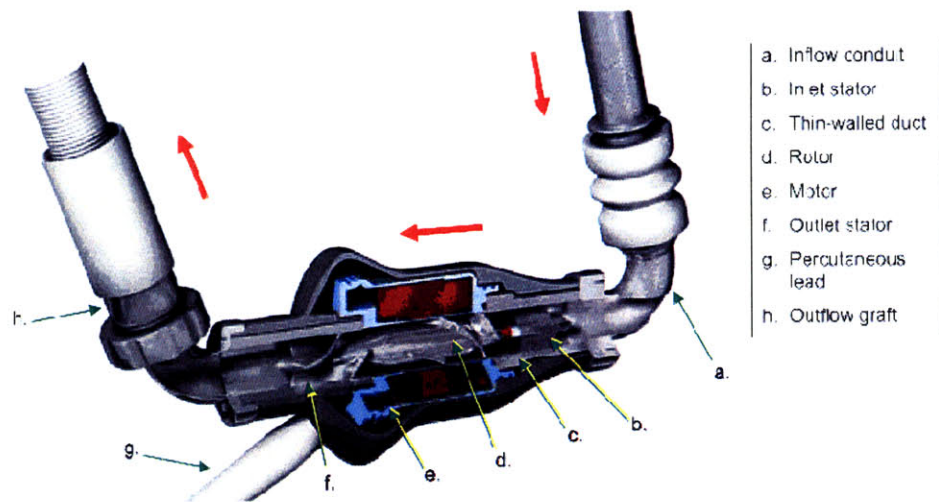
a. Inflow conduit
b. Inlet stator
c. Thin-walled duct
d. Rotor
e. Motor
f. Outlet stator
g. Percutaneous lead
h. Outflow graft

Figure 6 [17]. Components of HM II. HM II is a rotary pump that has a rotor that is rotated by the electromagnetic forces generated by the motor.

2.8.2.   HeartMate® III

The HM III is a continuous-flow, centrifugal-type rotary blood pump, featuring a bearingless motor with a magnetically levitated impeller, as shown in Figure 7. While slightly larger than the HM II, this device combines the functions of impeller rotation and levitation in a single magnetic structure. The pump is completely sheathed in titanium [20]. The device operates at a constant speed which can be adjusted though a monitor.
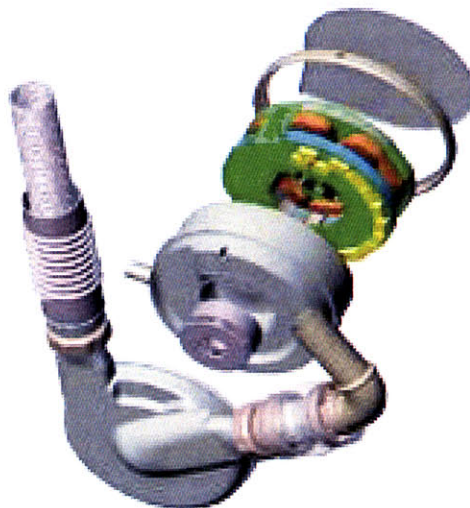

Figure 7 [20]. Exploded view of the HM III. This third generation LVAD features a bearingless motor with a magnetically levitated impeller.

2.8.3. CentriMag®

The CM is a continuous-flow, centrifugal-type rotary blood pump featuring a bearingless motor with a magnetically levitated impeller that is placed outside the body, as shown in Figure 8. The pump housing and rotor are made of medical-grade

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polycarbonate, and are designed for single-use purposes. The centrifugal pump design permits rotation of the impeller at lower speeds, while still achieving desired flow rates [21]. Figure 7 is a picture of the pump and its monitoring unit.

Figure 8 [22]. CM placed extracorporeally. The CM is a centrifugal-type rotary blood pump that is placed outside the body.

Figure 9 [22]. Picture of Centrimag®.

3. OBJECTIVE

The purpose of this study is to determine the cross-correlation between an aqueous DRP solution and blood pump hemolytic potential due to applied shear forces and/or heat. Polyethylene glycol (PEO), with a molecular weight of 5 kDa will be used as the drag reducing polymer in this experiment, as it has been shown be an effective DRP. If a cross-correlation exists between the degradation rate and hemolytic potential exists, then PEO can be used to assist the design of biomechanical devices by serving as a simple indicator of hemolytic potential that can be employed at any stage of the design process.

4. TEST PROCEDURE

4.1. *Test Design and Rationale*

The CM will be used to reproduce previous results and to verify the test set-up. Viscosity data and % drag reduction (%DR) will be measured to determine the degradation rate of the aqueous DRP solution. The same data will be taken for two biomechanical blood pumps HM II and HM III, products of Thoratec Corporation. This

data will be correlated with hemolysis data taken from previous bovine blood hemolysis studies to determine the predictive ability of the DRP for hemolytic potential.

4.2. *Experimental Setup*

To model the blood pump loop for *in vitro* testing, the device is connected to a reservoir. There are three variables in this loop setup: (1) pump speed, (2) flow rate, and (3) pressure gradient. The pump speed does not change unless it is manually altered. The flow rate and pressure gradient can be adjusted via the clamp. All three variables are monitored in the experimental setup. Figure 10 shows a schematic of the loop setup. The pump will either be the CM, HM II, or HM III.
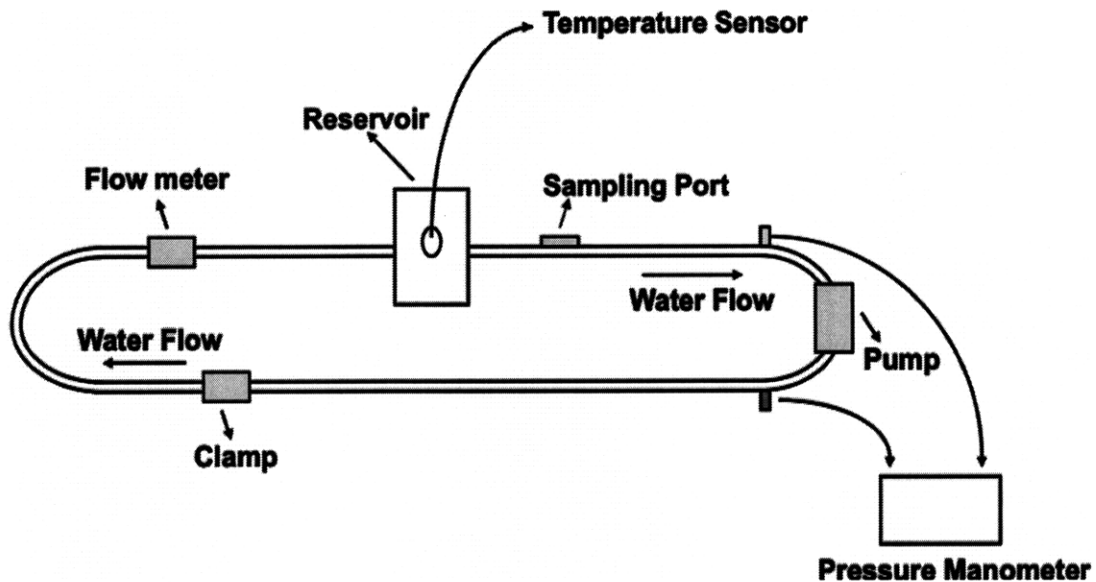

Figure 10: Schematic of loop setup. To model the loop inside the human body, the device is attached to a reservoir, which acts like the heart.

4.3. Procedure

4.3.1. *In-house testing using PEO*

4.3.1.1. *Sensitivity of PEO to temperature*

Heat generated by the pump in the loop heats the solution to a higher temperature. As discussed before, temperature affects the viscosity of the fluid. Therefore, to determine the sensitivity of the DRP to temperature, a temperature vs. viscosity curve needed to be generated. To do so, the Pitot tube was placed in a temperature controlled water bath, as shown in Figure 11. Since the water bath has a much larger fluid volume (approximately 2 L) than that which was placed into the Pitot tube (approximately 5 mL), it can be assumed that the temperature of the fluid would be equal to the temperature of the water bath when placed into the Pitot tube. Then the temperature was varied and the polymer's viscosity was determined.
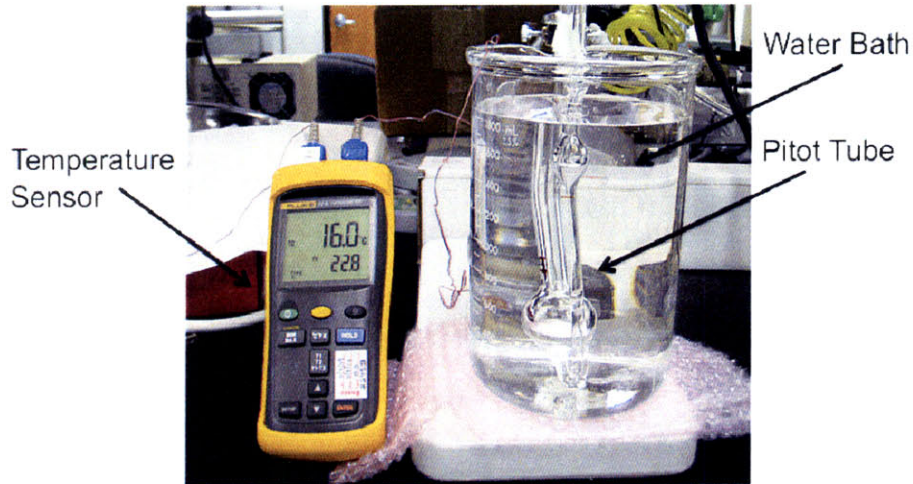
13

Figure 11: Determining the sensitivity of PEO to temperature. The Pitot tube was placed into a temperature-controlled water bath and the viscosity of the DRP was measured at different solution temperatures.

4.3.1.2.   *Verification of PEO as a DRP using CentriMag*

A previous study, conducted by Sobajima, et al., investigated the mechanical degradation of DRPs to determine whether the polymer can act as a potential hemolytic indicator for blood pumps. This study used 1 L of PEO (1000 ppm, 4500 kDa) and maintained a constant pressure gradient of 300 mmHg. The solution was driven through a centrifugal pump, either a Bio Pump (Medtonric, Inc.) or CentriMag (Levitronix, Inc.). Polymer degradation was evaluated through reduction in both the solution flow rate and molecular weight for two hours.

The previous study was replicated, using the same polymer concentration and constant pressure gradient, to verify the behavior of PEO as a DRP and a potential hemolytic indicator. The pump speed was set to 3800 rpm, which when using blood analog, produced a pressure 300 mmHg and a flow rate at the design point of 5 L/m. Though the previous study using PEO with a molecular weight of 4500 kDa, a molecular weight of 5000 kDa was used instead.

4.3.1.3.   *In-house hemolysis testing*

For HM II and HM III, a similar setup, as shown in Figure 10, was used. As before, the concentration of PEO was 1000 ppm, with the polymer having a molecular weight of 5000 kDa. For the two blood pumps used in this experiment, the pressure gradient was maintained at 100 ± 3 mmHg, as this matches the ASTM standard (F 1841-97).

The pump speed was determined using head pressure (H) vs. volumetric flow rate (Q) curves, as seen in Figure 12 and 13. These curves are generated using a blood analog (i.e. a solution with the same viscosity and density as blood) and a similar flow loop with a flow limiting clamp. The beginning of each curve is determined when the clamp is tightened to the point where there is no fluid flow or 'pump shut-off'. The curves end

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when the clamp is fully open allowing the maximum fluid flow at that pump speed. HQ curves are specific to a specific pump design, much like fingerprints. For a given pressure and flow rate, it is possible to determine the pump speed at which the device should run at. For the HM II, in order to obtain a pressure of 100 mmHg and the design point of 6 L/min, the pump speed was 11,000 rpm. For HM III, to obtain a pressure of 100 mmHg and a design point of 6 L/min, the pump speed was approximately 3750 rpm. With the pump speed and pressure set, the flow rate was monitored through 4-hour long experiments. The change in viscosity over time as well as the %DR was calculated.
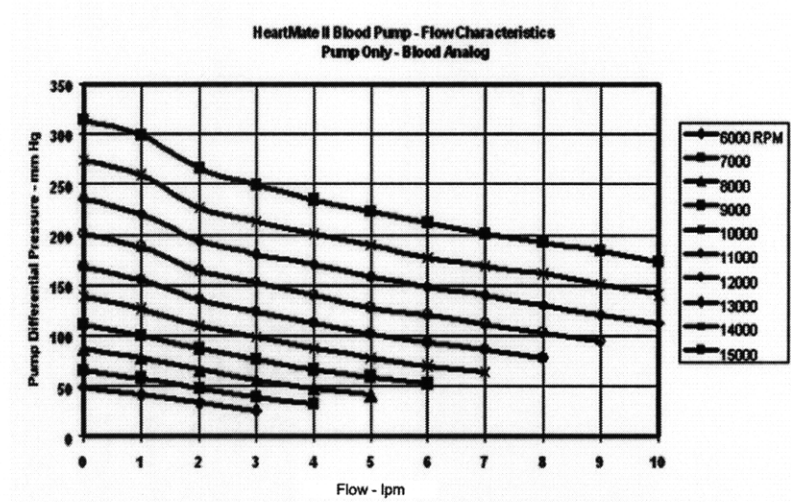

Figure 12. Flow Characteristics for HM II (HQ Curve). This HQ curve will be used to determine the speed at which the pump will need to run at in order to achieve 6l/min flow at 100 mmHg pump head.
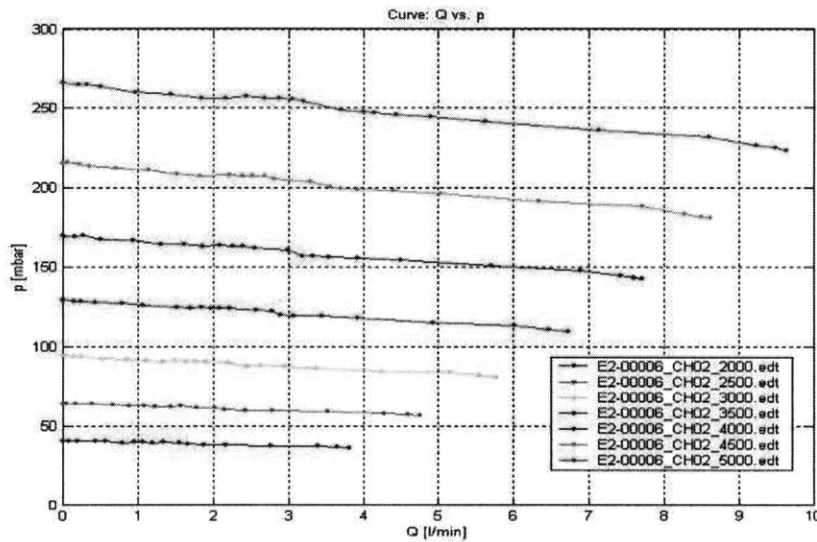

Figure 13 [23]. Flow characteristics of HM III. This HQ curve will be used to determine the speed at which the pump will need to run at in order to achieve 6l/min flow at 100 mmHg pump head.

In this experiment, the percent flow reduction (%FR) is defined to be the percent at which the DRP's flow rate is reduced below that of blood analog's flow rate at a given pump speed and head, as shown in Equation 3.

$$\% FR = \frac{Q_{B.A.} - Q_{DRP}}{Q_{B.A.}} \times 100 \qquad (3)$$

To create a 1000 ppm solution, PEO is mixed with water using a magnetic stirrer. When the solution was fully dissolved, it is placed into the experimental loop. Samples are taken every twenty minutes for the first two hours, and then every half hour for the last two hours. The solution's viscosity is measured using a Pitot tube. The temperature change of the solution is monitored with each sample. At least three experiments of each type of heart pump will be conducted.

5. RESULTS

5.1. Sensitivity of PEO to temperature

The relationship between viscosity and temperature of a 1000 ppm PEO solution is inversely linear, as shown in Figure 14. As predicted, as temperature increases, viscosity decreases. The best-fit line (shown as the solid black line) is

$$y = -0.606x + 4.1734 \qquad , \qquad (4)$$

where x is the temperature [Celsius] of the solution and y is the viscosity [cP] for that given temperature.
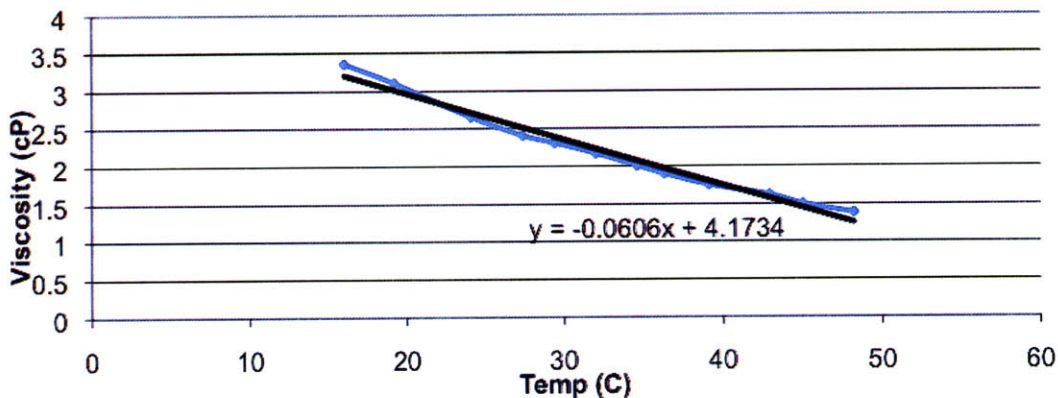

Figure 14. Sensitivity of PEO's viscosity to temperature. Results showed that for every Celsius degree increase, the viscosity decreased by 6.06%.

Using the best-fit line, it can be seen that for every degree Celsius increase, there is a 6.06% decrease in viscosity. Therefore, to take into account the effect of temperature on viscosity, the viscosity measurements will be normalized use Equation 4.

5.2. Verification of PEO as a DRP in CentriMag®

Results obtained in the verification of PEO as a DRP were very similar to results obtained from the previous study. Results from the previous study can be found in Figure 16a and 16b. Drag reduction began at approximately 30% and ended at around 10%. The change in drag reduction, from this study, can be found in Figure 17 and the change in viscosity over the two hour experiment can be found in Figure 18.
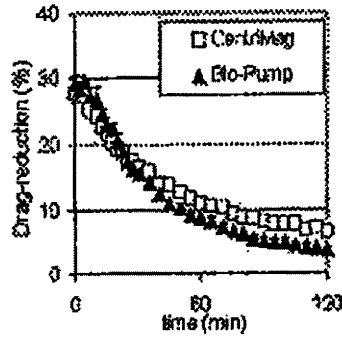

Figure 16 [1]. Previous results using the CentriMag®. The x-axis is a function of time and the y-axis is a fnction of drag reduction (%). Results showed that the longer the polymer was placed into the CentriMag®, the more drag reduction occurs.
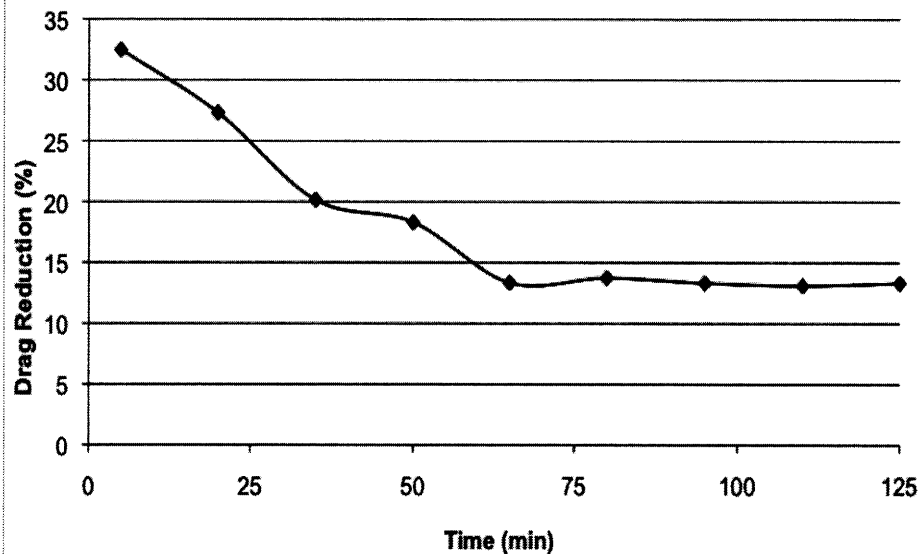

Figure 16: Drag reduction of CM. The verification of PEO as a DRP was successful, with the drag reduction curve being very similar to that which was produced by a previous study [1]. As time increases, drag reduction also increases.
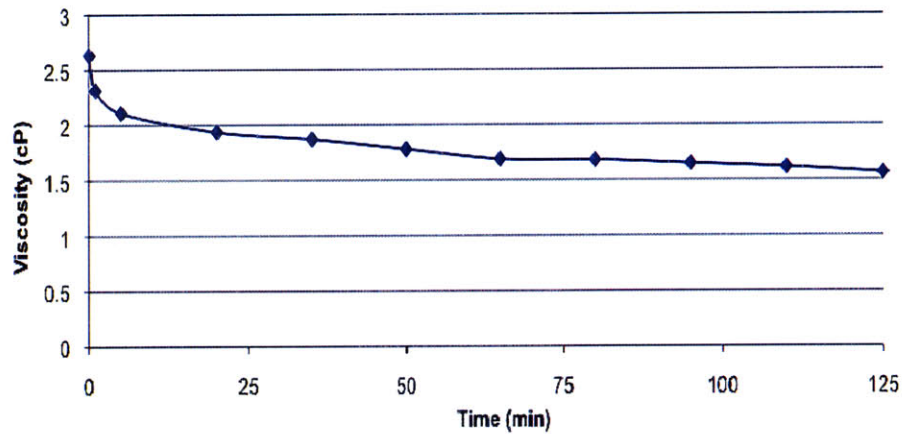
Figure 17: Change in viscosity of PEO over two hour period in CM. This curve shows that the polymer degraded over time, due to shearing within the CM.

5.3. HeartMate® II

Four experiments were conducted using an aqueous solution of the polymer. The HM II loop setup can be found in Figure 18.
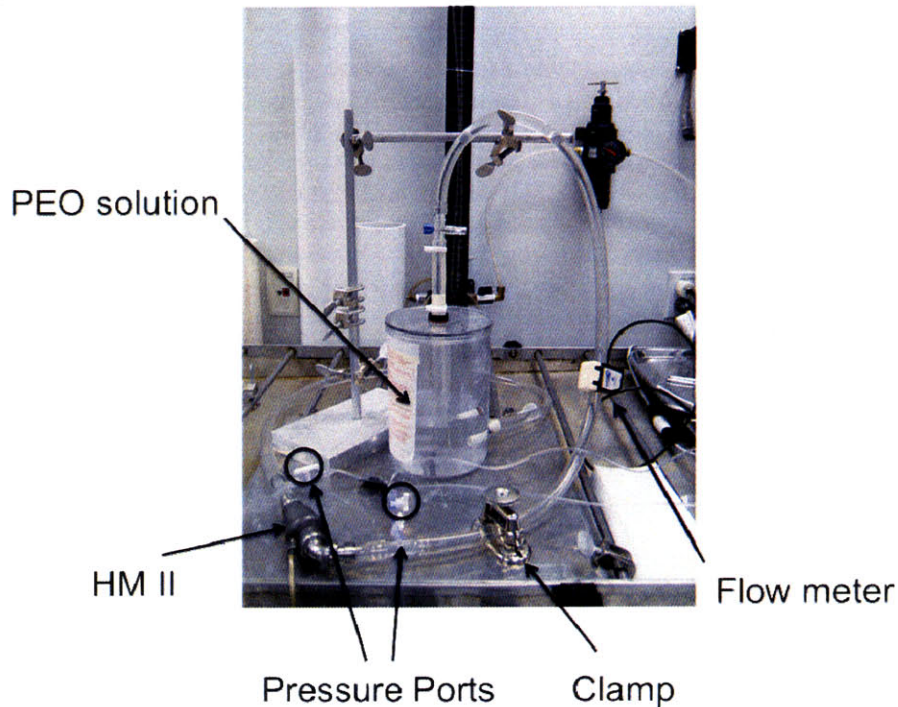

Figure 18: HM II loop setup. The loop setup includes a flow meter and pressure ports. Not pictured, though monitored, is the temperature probe.

The average initial normalized viscosity was 2.78 cP and the steady-state viscosity was, on average, 2.57 cP. The average initial %DR was 98.5% and the final %DR was 10.7%. Graphs of the individual experimental results can be found in Figure 19 and 20.
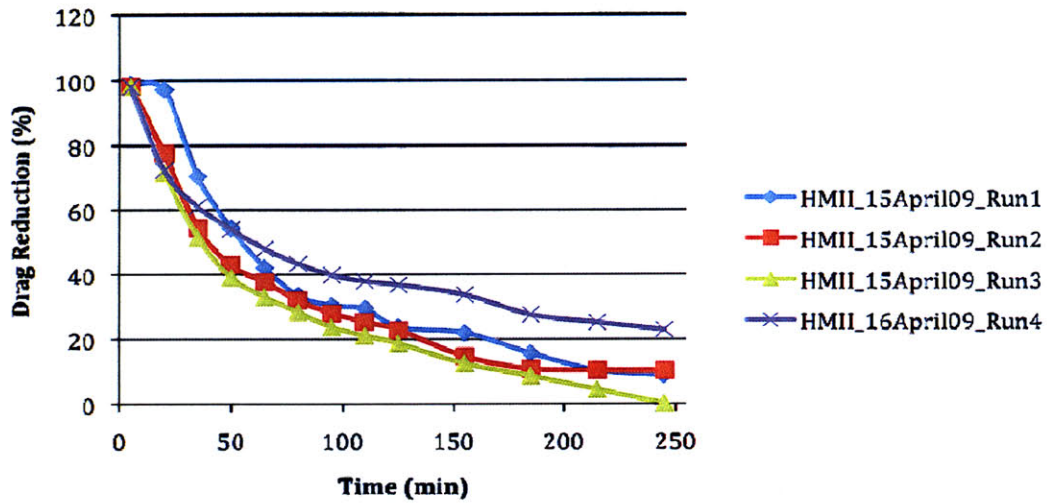
Figure 19. Percent drag reduction for HM II. The average initial %DR was 98.5% and the final %DR was 10.7%.
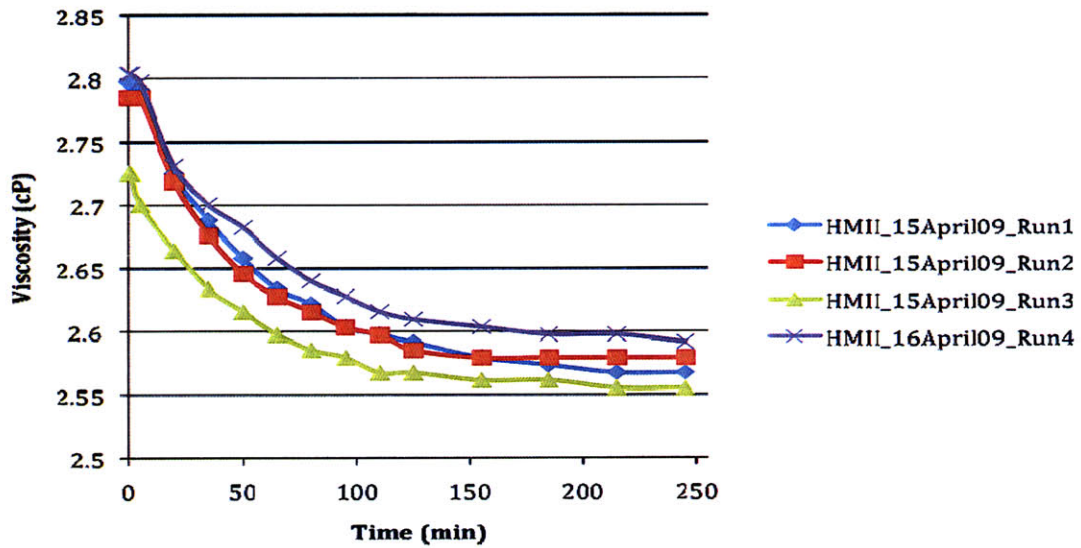

Figure 20. Change in viscosity for HM II. This graph shows the change in viscosity of PEO in four hour experiments. The average initial normalized viscosity was 2.78 cP and the steady-state viscosity was, on average, 2.57 cP.

5.4. HeartMate® III

Two experiments were conducted using an aqueous solution of the polymer with HM III. The HM III loop setup can be found in Figure 21.
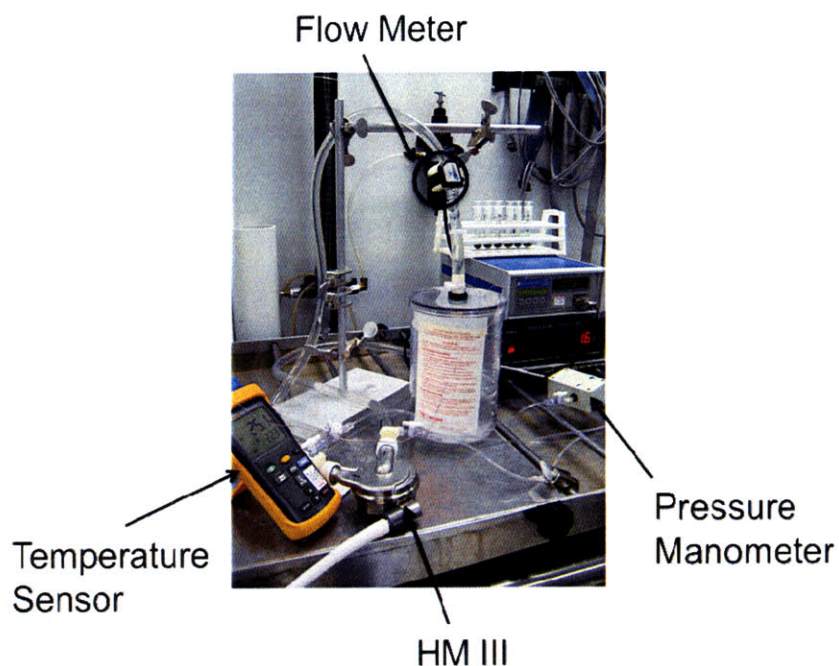
Figure 21. Experimental setup for HM III. The loop setup includes a flow meter, temperatures sensor, and pressure manometer.

The average initial viscosity was 2.81 cP and the final average viscosity was 2.65 cP. All viscosity measurements were normalized to the temperature of the fluid. The average initial %DR was 70.6% and the final average %DR was 12.5%. For Runs 2 and 3, at approximately twenty minutes into the experiments, the flow rate of the solution was higher than flow rate of blood analog (assuming the same pump speed and pressure); this resulted in a negative %DR. The individual experimental results for percent drag reduction and changes in viscosity can be found in Figure 22 and 23, respectively.
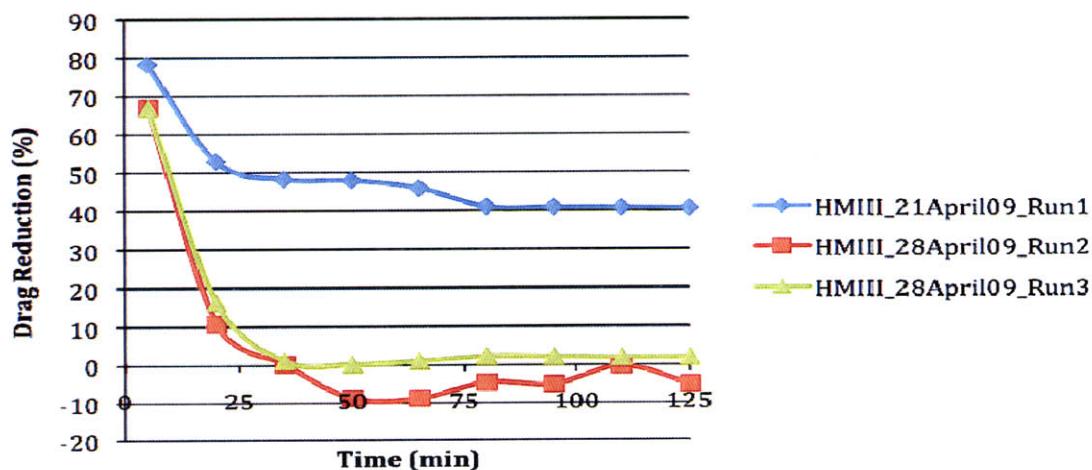

Figure 22. Percet drag reduction for PEO in HM III. The average initial %DR was 70.6% and the final average %DR was 12.5%.
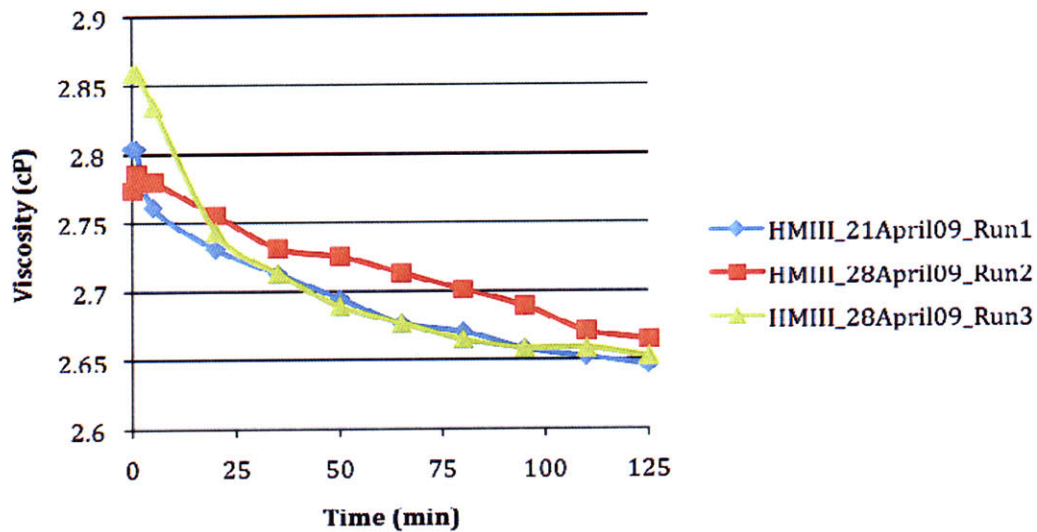
Figure 23. Change in viscosity over time for HM III. The average initial viscosity was 2.81 cP and the final average viscosity was 2.65 cP.

6. DISCUSSION

The results of 28 hemolysis tests of the Thermo Cardiosystems HM III model F1d, a PM- (powdered metallurgy) textured titanium prototype showed that the average hemolysis rate was 1.38 ± 0.63 g/day and the milligram normalized index of hemolysis (mg N.I.H.) was 0.571 ± 0.333. In that same study, the results of four tests of a HMII device, having smooth titanium blood-contacting surfaces, showed that the average hemolysis rates was 3.80 ± 1.11 g/day and the milligram N.I.H. was 0.0393 ± 0.0155 [24]. These values indicate that HM II can cause more hemolysis than HM III.

Comparing the results of the current study, HM II has a larger mathematical difference between the average final and average initial viscosity. The difference for HM II, with a total experimental time of 250 minutes, was 0.21 cP, while the difference for HM III, with a total experimental time of 125 minutes, was 0.16 cP. If the mathematical difference is larger, this is an indication that more shearing has occurred, which would results in larger initial and final viscosities. The experiment for each device was terminated when the final viscosity reached a steady state value. Comparing both devices at 125 minutes, HM II had a change in viscosity of .19 cP while HM III had a change in viscosity of 0.16 cP. While the difference is narrowed, the change in viscosity for HeartMate II is still seen to be greater than that of HeartMate III. This indicates that HeartMate II has a shearing rate larger than that of HeartMate III.

The positive correlation between using the polymer and actual hemolysis results, shows that in early stages of design, if an aqueous solution of PEO is placed in the biomechanical device, and the difference between the final and initial viscosity is within the boundary of 0.16 cP and 0.21 cP, then the biomechanical device should have an acceptable hemolysis rate because both HM II and HM III have acceptable hemolysis rates. Additionally, if the viscosity difference is below 0.16 cP, then the design should also acceptable due to the fact that a lower difference would indicate less shearing and lower hemolysis rate. However, if the viscosity difference is significantly greater than 0.21 cP, then redesigning the LVAD may be necessary.

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Variability between the initial viscosity, which would then affect the initial flow rate, may be due to the time length allowed for mixing the polymer into its aqueous form. It was noticed that if the solution was able to mix for longer periods of time, such as for a few days, the initial viscosity would be lower than if the solution was allow to mix overnight (approximately 20 hours). This behavior indicates that an aqueous solution of PEO is a non-Newtonian fluid and acts like a shear-thinning fluid, where adding stress decreases the viscosity. Simultaneously, this alludes to the question of whether the polymer was fully mixed into the solution when placed into the loop. Based on visualization, when the polymer or clumps of the polymer were no longer visible to the naked eye, the solution was assumed to be fully diluted. Additionally, it was noticed that if the solution was simply left overnight, without any mixing or any stress, the viscosity decreased over time. However, when the solution was mixed again after being fully diluted, the viscosity increased back to its initial viscosity.

Looking at the drag reduction graph for HM III (Figure 15), there seems to be a significant difference between the first curve and the second and third curves. The latter two runs result in a flow rate that is higher than the flow rate of blood analog under the same pump settings. However, the final viscosities for all three runs are approximately equal to 2.67 cP. The final viscosity of the polymer in HM III is less than the viscosity of blood analog, which is 3.6 cP. This is interesting to note and additional testing should be conducted.

The results of this experiment indicate that many questions remain regarding PEO's behavior under shear stress and simple mixing. While this research shows the positive correlation of the experimental results and hemolysis results, more research is needed to fully understand PEO and it's potential to be a hemolytic potential indicator for designing LVADs and other biomechanical devices.

7. CONCLUSION

The result of this experiment showed that PEO has a positive potential to be a hemolytic indicator in the early stages of biomechanical design. The mathematical difference between the average final and initial viscosities for HM II was greater than the difference for HM III. HM II had a mathematical difference of 0.21 cP and HM III had a 0.16 cP. This corresponds to HM II having a larger mg. N.I.H. and a larger average hemolysis rate. HM II had an average hemolysis rate of 3.80 ± 1.11 g/day and the mg. N.I.H. was 0.0393 ± 0.0155. HM III had an average hemolysis rate of 1.38 ± 0.63 g/day and the milligram normalized index of hemolysis (mg N.I.H.) was 0.571 ± 0.333.

The positive correlation between using the polymer and actual hemolysis results, shows that PEO can be a good indicator of hemolysis. If the difference between the final and initial viscosity of the aqueous solution is less than 0.21 cP, then hemolysis rate should be acceptable. However, if the difference is much higher than 0.21 cP, it is recommended that changes to the design be considered.

While there is a positive correlation, addition research must be conducted to further understand the molecular behavior and characteristics of PEO. While the boundary range may be 0.16 cP and 0.21 cP for these two devices, these boundary values may be significantly different for other types of biomechanical devices or even differently designed LAVDs. Therefore, additionally testing should be conducted using a large range of biomechanical devices, to continue to determine the hemolytic potential of PEO.

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