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SANTA CLARA UNIVERSITY

Department of Bioengineering

I HEREBY RECOMMEND THAT THE THESIS PREPARED
UNDER MY SUPERVISION BY

Joseph Anthony Ikaika Choy, Nicholas Domek, John 'Patrick' Tavelli

ENTITLED

INNOVATIONS IN TRAUMATIC HEMORRHAGE

BE ACCEPTED IN PARTIAL FULFILLMENT OF THE REQUIREMENTS
FOR THE DEGREE OF

BACHELOR OF SCIENCE
IN
BIOENGINEERING

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INNOVATIONS IN TRAUMATIC HEMORRHAGE

By

Joseph Anthony Ikaika Choy, Nicholas Domek, John “Patrick” Tavelli

SENIOR DESIGN PROJECT REPORT

Submitted to
The Department of Bioengineering

Of

SANTA CLARA UNIVERSITY

In partial fulfillment of the Requirements
for the degree of
Bachelor of Science in Bioengineering

Santa Clara, CA

2016

INNOVATIONS IN TRAUMATIC HEMORRHAGE

Joseph Anthony Ikaika Choy, Nicholas Domek,
John “Patrick” Tavelli
January 29, 2016

Abstract

Traumatic hemorrhagic injuries present a great problem to humanity and a challenge to medicine in the modern world. Current methods of treating these injuries in the field are ineffective and often extremely overkill or injurious. These methods are particularly inadequate when applied to the continuous high pressure bleeding that occurs from arterial wounds. Our project focuses on lowering the barriers to entry to innovation in the field of bleeding treatment by creating a low cost model of the human circulatory system. This model can function as a low-cost testing platform for novel bleeding treatments developed by companies and individuals that do not have the resources to regularly purchase extremely expensive cardiovascular simulators. To this end we designed a tripartite model which included a heart-simulating pump, vessel-simulating vasculature, and blood-mimicking fluid. In order to ensure our device functioned as a testing platform, we performed some preliminary solution candidate tests on it which had the ancillary benefit of identifying one effective but biologically unsafe solution that could be translated into a safe and efficacious future solution. Ultimately we found that our system functioned well as a testing platform for traumatic injury treatments and that standard silicone sealant administered by injection into the vessels had the greatest efficacy in stopping bleeding.

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1.1 Background

1.1.1 The Problem

The title of this Senior Design Thesis, Innovations in Traumatic Hemorrhage, identifies the primary problem it investigates: hemorrhage. Trauma is the rather broad medical term for "An injury (as a wound) to living tissue caused by an extrinsic agent."[1] Hemorrhage is the formal medical term for "bleeding," the condition wherein the human cardiovascular system suffers an unintentional loss of blood to the outside world – usually through the compromise of a blood vessel wall. This thesis, then, seeks to investigate and propose novel treatment solutions for bleeding as caused by extrinsic forces.

1.1.1.1 The Effects of Hemorrhage

It is easy to grasp the problematic nature and gravity of traumatic hemorrhage. Adequate circulation of blood throughout the body is a necessary precondition for life. Without a steady supply of oxygen and the continuous removal of carbon dioxide, living cells begin to die in minutes.[2] This condition is referred to as "shock." If the cause of the shock is a loss of blood, it is specifically called "hypovolemic" (too little volume) shock. By way of hypovolemic shock, a loss of blood can rapidly and dangerously escalate into a loss of life.

The human cardiovascular system, on average, circulates between 3.5 and 6 liters of whole blood every minute.[3] The flow rates required to accomplish this will be discussed in some detail later, but suffice to say that it does not take long to drain the system if there is a bad leak. Bleeding is ubiquitous – what child hasn't scraped their knee or papercut their finger? But bleeding is also dangerous and fast, and thousands die every year as a result of it.

1.1.1.2 Hemorrhage Moieties

[1] Gove, Philip Babcock. *Webster's Third New International Dictionary of the English Language, Unabridged: A Merriam-Webster*. Springfield, MA: G. & C. Merriam, 1961.
[2] "Questions and Answers About Transplantation and Donation." University of Michigan Transplant Center: TransWeb.org. February 4, 2010. Accessed January 10, 2016. http://www.transweb.org/faq/q3.shtml.
[3] United States of America. Department of Energy. Oak Ridge National Laboratory. *Reference Values for Total Blood Volume and Cardiac Output in Humans.* By Lynn R. Williams. Oak Ridge: Martin Marietta Energy Systems, 1994.

The broader injury category of "hemorrhage" can be subdivided into several distinct moieties. The first way in which bleeding is categorized has to do with its path of exit from the cardiovascular system. *Internal* bleeding describes errant flow that exits the cardiovascular subsystem into a different closed space or system: generally into body cavities or larger organs. Internal bleeding is often caused by blunt force impacts such as are incurred in falls, collisions with larger objects, or post-explosion shockwaves. *External* bleeding is characterized by undesired blood flow from the cardiovascular system into the outside world. This typically results from high pressure, low impact area injuries like punctures or gunshots.

Internal and external hemorrhage can be further subdivided by the components of the cardiovascular system that have been breached. Capillary hemorrhage is the least concerning category. Scraped knees and papercuts could be described as external capillary hemorrhage. Venous bleeding is more serious – the majority of the blood in the body is stored in the veins, albeit at low pressure. Thus venous bleeding is concerning and dangerous, but controllable. Arterial bleeding is the most dangerous of all, characterized by high pressure, irregular flow that is difficult to control and extremely deadly.

1.1.1.3 The Scope of the Problem and Affected Populations

This project is concerned primarily with treatments for internal and external traumatic, arterial hemorrhage. These are the most dangerous and least predictable forms of bleeding, and they result in the most deaths. The National Trauma Institute estimates that approximately 180,000 individuals die every year as a result of traumatic injury, of which 30-40% (72,000 deaths) can be directly attributed to hypovolemic shock and hemorrhage.[4,5] This statistic only known deaths which occurred in the United States – the global figure is much, much larger. Figure 1.1 below provides a good summary of these numbers.

Extrapolating from the global population of around 7,300,000,000 it is not unreasonable to expect that around 1,600,000 people die annually due to traumatic hemorrhage.[6] This is a staggeringly large figure. In fact, the only traumatic injury with a higher global morbidity rate is neurologic trauma.[7] These numbers become even more staggering in certain contexts. One particularly notable example is within the military, where more than 80% of deaths are hemorrhage related.[8] The rates are also higher in developing nations, where medicine is not as advanced, and in rural environs. It is on these three areas that we focus in this thesis:

[4] Choy, Joseph A., Nicholas Domek, and John P. Tavelli. "Tissue Engineering in Traumatic Injury and Blood Loss." Final Report: Bioengineering 172 – Tissue Engineering, Santa Clara University, 2014.
[5] Trauma Statistics. (2014, February 1). *National Institute of Health* (November 30, 2014)
[6] U.S. and World Population Clock. (2016, January 10). Retrieved January 10, 2016, from the United States Census Bureau, http://www.census.gov/popclock/
[7] Kauvar, David S., Rolf Lefering, and Charles E. Wade. "Impact of Hemorrhage on Trauma Outcome: An Overview of Epidemiology, Clinical Presentations, and Therapeutic Considerations." *The Journal of Trauma: Injury, Infection, and Critical Care*, 60, no. 6 Supplement (June 2006): S3-S11.
[8] Katzenell, Udi, Nachman Ash, Ana L. Tapia, Gadi A. Campino, and Elon Glassberg. "Analysis of the Causes of Death of Casualties in Field Military Setting." *Military Medicine* 177, no. 9 (September 2012): 1065-068.

1. Military Emergency Medical Services
2. Developing World Healthcare
3. Rural Emergency Medical Services

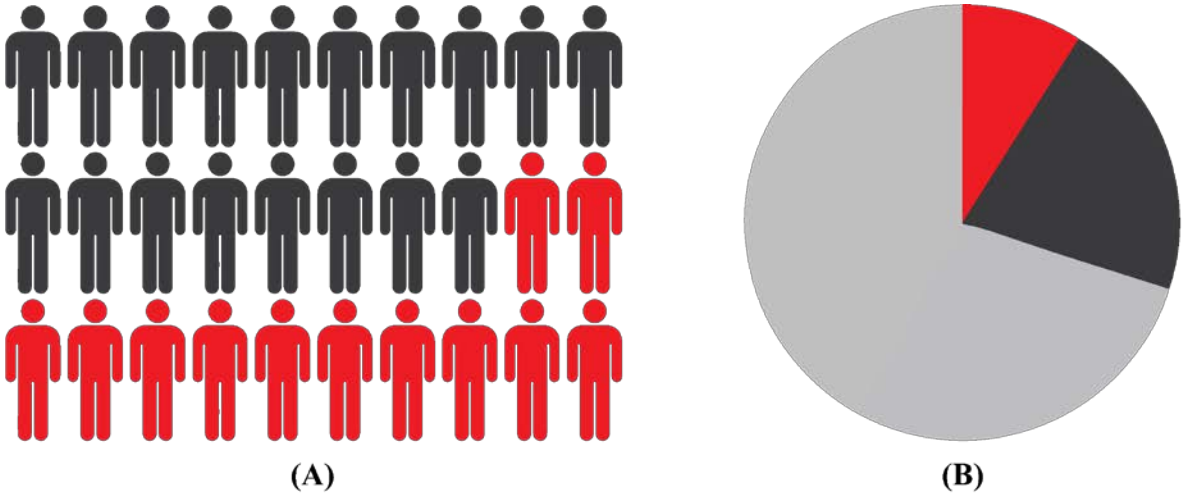


Figure 1.1 Lives and Life Years Lost to Traumatic Hemorrhage
(A) Each stick figure represents 6,000 lives lost annually in the United States as a result of traumatic injury. Red figures represent lives lost directly to hemorrhage. (B) Graph represents total life years lost in the United States annually to illness or trauma. The red segment indicates life years lost to hemorrhagic trauma, the dark gray segment indicates life years lost to other traumatic injuries, and the light gray segment represents medical life years lost.

1.1.1.4 Summary and Importance of the Problem

These statistics and realities stem from several key facts about traumatic hemorrhage and its treatment in the world today. These are listed below:

1. The number of functional “in-the-field” treatments is limited
2. Existing “in the field” and “in hospital” treatments are limited in their effectiveness, particularly when it comes to internal bleeding
3. Access to “in hospital” treatments is severely limited, particular in the populations that are identified above

What current traumatic blood loss prevention methods exist are aimed mainly at temporary bleeding control and do little to protect the injury site itself or encourage eventual repair. They are also exclusively limited to injuries located on the extremities and provide no solution for arterial bleeding or bleeding anywhere in the thoracic region. Tissue engineering aimed at the design of better treatments for traumatic injury and prevention of blood loss is greatly needed.

Traumatic hemorrhagic injuries present a great problem to humanity and a challenge to medicine in the modern world. Current methods of treating these injuries in the field are ineffective and often extremely overkill or injurious. These methods are particularly inadequate when applied to the continuous high pressure bleeding that occurs from arterial wounds. Our project focuses on stopping this kind of arterial hemorrhage with an injectable material that can be easily and effectively applied anywhere.

1.2 Review of Field Literature

1.2.1 Treatment of Traumatic Hemorrhage

The first phase of our literature review focused on current and emerging treatments for traumatic hemorrhage. There are hundreds of effective treatments for traumatic hemorrhage available ranging from the classic: pressure, gauze, and tourniquets; to the novel: hemostatic polymeric gels, fibrin glues, expandable micro sponges, and more. These solutions can be broadly categorized into field and clinical settings, with our project's particular focus being field treatments. These solutions include both mechanical and biological treatments that offer unique solutions to different traumatic bleeding situations. Some current solutions are shown in Figure 1.2 immediately below.

(A) (B) (C)

Figure 1.2 Various Current Approaches to Hemorrhage Control
(A) The classic method of extreme field hemorrhage control – the tourniquet (B) The classic method of extreme clinical hemorrhage control – surgical intervention (C) A new approach to field hemorrhage control – the recently FDA approved XStat device – which stops axial bleeding injuries in seconds

1.2.1.1 EMS Field Approaches

Current basic field treatment of traumatic bleeding injuries revolves around an escalating three step regimen. First, pressure is applied outside the wound with gauze, towel, or another form of compress. In the second step, pressure is shifted from outside the wound to inside the wound as close to the bleeding site as possible to maximize contact surface area. In the third step, a tourniquet is applied (if possible) as close as possible to the site of injury to stop blood flow distal to the application location. During these steps, other methods of bleeding control are often conducted, including elevation and pressure at body pressure points. This method of bleeding control is reliable, inexpensive, and well-tested, but it does not apply to all injuries and may not be fast enough to prevent a patient from going into hypovolemic shock before getting to a hospital. To address these and other concerns, new traumatic bleeding control technologies have emerged to give field responders an expanded toolkit for saving lives.

1.2.1.2 Military Technology

Technologies targeted toward the military market have made significant progress in increasing the speed and efficacy of traumatic injury treatments. One such technology currently in use by the United States Armed Forces is QuickClot Combat Gauze ® which is a fabric impregnated with zeolite, a hemostatic agent which can decrease the time to stop bleeding and promote coagulation, which has been shown to increase chances of survival in injured soldiers.[9] Another treatment technology developed by RevMedX called the X-Stat utilizes a syringe injection of a multitude of micro-sponges to provide internal hemostatic pressure to traumatic injuries in the groin and axilla regions that decreases blood loss and increases clotting efficiency.[10] This technology provides treatment to areas of the body that previously could not be treated with pressure. The sponges have a radiopaque marker on them to assist in removal in a clinical setting after bleeding control.

1.2.1.3 Gels, Foams, and Glues

Modern researchers have produced polymer gels, foams, glues that offer attractive treatment solutions for incompressible external traumatic bleeding injuries by filling the site and speeding up the natural clotting process. Suneris has invented a plant based injectable polymer gel called Vetigel that adheres to the surrounding tissue, promotes platelet accumulation, and can halt 2x arterial pressure in seconds. Chitosan and gelatin based foams have also been developed to induce rapid hemostasis on traumatic injury sites upon application to the injury surface. One novel two-part foam can even be injected into the abdominal cavity to provide hemostatic pressure to slow or stop internal bleeding of the vital organs. Fibrin glues offer a similar solution through a different pathway. Fibrin glue works by combining thrombin and fibrinogen to instantly form a natural clot instantaneously without having to wait for the natural coagulation process.

1.2.1.4 Biological Treatments

In addition to the previously discussed technologies for treating external bleeding, there are also a number of biological treatments in the research phase for treating internal injuries. Two areas of research were of particular interest to our project: platelet-like nanoparticles and a synthetic fibrin cross-linking polymer. Researchers at the University of California at Santa Barbara invented nature inspired platelet-like-nanoparticles (PLN's) that mimics the shape, flexibility, and surface chemistry of a platelet and could be used to supplement the body's own supply of platelets and also follow the same basic pathway of platelets to induce clot formation.[11] By mimicking the shape and flexibility of platelet the PLN's are able to bind to the site of injury and induce clotting at that specific site. Platelets do not cause clots in random areas of the body and neither would the PLN's. Following binding to the injury site, the PLN's would then call other

[9] Gegel, Brian T., Austin, Paul N., Johnson, Arthur. "An Evidence Based Review of the Use of a Combat Gauze (QuickClot) for Hemorrhage Control." *AANA Journal* 81, no. 6 (December 2013). 453-458.
[10] "De Novo Classification Request for XSTAT." Silver Spring, Maryland: FDA, United States Department of Health and Human Services (January 28, 2013). K130218.
[11] Fernandez, Sonia. "Bio-Inspired Bleeding Control." *UC Santa Barbara Current*, Santa Barbara, CA: University of California, Santa Barbara (November 12, 2014). Web. <http://www.news.ucsb.edu/2014/014506/bio-inspired-bleeding-control>

platelets to the site of the injury using factors attached to its surface this mimics the same mechanism of the body's own platelets.

The second internal bleeding technology we looked at was a synthetic fibrin cross-linking polymer developed by researchers at the University of Washington. The polymer, named PolySTAT, acts similarly to transglutaminase factor XIII in the blood, which is a cross linking molecule to increase the stability of a fibrin clot, except this polymer increases both the fibrin binding efficiency and strength of the clot.[12] The polymer will circulate in the blood until it comes into contact with the fibrin network, and it is bio-resorbable with organic by-products. This polymer offers great potential for an internal bleeding treatment, especially in patients on blood-thinners or with coagulopathy, but extensive research is still needed to test the potential toxicity and carcinogenicity of PolySTAT.

1.2.1.5 Clinical Approaches

The most common treatment of traumatic bleeding in a clinical setting is surgery. This involves the surgeon using sutures or a heat probe to stitch or cauterize the broken blood vessels in the injury site. Surgery is very specific and can be applied to both internal and external bleeding because advanced imaging technologies are available (CT, MRI) to locate the exact site of bleeding and fix the problem. Internal bleeding can be difficult to diagnose because tissues and organs are capable of absorbing significant amounts of blood, especially if the injury has been bleeding for an extended period of time.[13] The embolization technique is another common clinical endovascular procedure to control traumatic bleeding.[14] An interventional radiologist will make an incision in the patient above the site of the injury and insert a catheter into the patient's blood vessel. Using camera imaging techniques, the radiologist guides the catheter through the patient's blood vessel or artery and find the site of bleeding. The catheter can then be used to facilitate the release of clotting factors.

1.2.2 Experimental Models

The second phase of our literature review focused on reviewing existing testing models for traumatic hemorrhage treatments. Experimental models are categorized into four distinct systems: mechanical models, computer models, animal models, and human models. Given the monetary, time, and ethical constraints of our research, animal and human models were not appropriate. However, we do touch on animal and clinical models to provide background and validity to the mechanical and computer models. Our primary focus for this particular project was mechanical models, which offer the advantages of being inexpensive, reliable, and easy to collect data.

1.2.2.1 Physiological Models

[12] Langston, Jennifer. "An Injectable UW Polymer Could Keep Soldiers, Trauma Patients from Bleeding to Death." *UW Today*, Seattle, WA: University of Washington (March 10, 2015). Web. <http://www.washington.edu/news/2015/03/10/an-injectable-uw-polymer-could-keep-soldiers-trauma-patients-from-bleeding-to-death/>

[13] Mistovich, Karen. *Prehospital Emergency Care*, ed. 10, Upper Saddle River, NY: Prentice Hall (July 7, 2013).

[14] Lopera JE. Embolization in trauma: principles and techniques. *Semin Intervent Radiol*. 2010;27:14–28.

Physiological models are the ideal testing model for traumatic bleeding injuries given the enormous complexity of the cardiovascular system. Many of the articles researched for this project tested the solution by inducing a traumatic bleeding injury to rats or mice then applying the given treatment to the animal and analyzing the immediate and long term effects. This can be effective at testing the viability of the solution, but it can be complex and difficult to gather quantitative data. Even more effective than animal models are human models. Due to the nature of traumatic bleeding, it is unethical, illegal, and impossible to develop controlled traumatic bleeding studies on humans. The main source for human models in this this area is patients being treated by civilian and military EMS, but the use of experimental treatments is extremely difficult.

1.2.2.2. Mechanical Cardiovascular Model

Mechanical models of the cardiovascular system do not include all of the elements of an ideal physiological model, but they offer the advantage of greater flexibility and control of the experiment. The pulsatile nature of the heart is essential for a mechanical model. Harvard Instruments has created a reliable, programmable pulsatile pump that can mimic the full blood flow patterns of the heart into a tubing system. Continuous flow pumps can also mimic blood flow, but move further away from the ideal physiological model. The academic research area of Particle Image Velocimetry (PIV) utilizes flow phantom models that do an exceptional job at mimicking the size, shape, and properties of arterial and venous vessels.[15]

There are a number of commercially available phantoms that allow researchers to researchers to create their own multi-branched tubing systems using molds and a two-part silicone polymer. Silicone tubing is the cheapest and most effective way to mimic blood vessels currently available. PIV studies also use viscous blood mimicking fluids (BMF), which is another essential element of a mechanical model of the circulatory system. These BMF's usually use a mixture of water, glycerol, dextran, and small polymeric microparticles sometimes made of silk.

There was additionally, in our research, one mechanical model constructed by another university's senior engineering class that had significant relevance to our project. The project, titled "Design of a Mechanical Model for Pulsatile Aortic Flow," was submitted as a senior thesis at Worcester Polytechnic Institute.[16] The students in question designed and built a cam-based peristaltic pump to induce pulsatile aortic flow through a network of blood vessel simulating Penrose tubes. The project attempted to recreate an aortic model so that experiments could be performed to determine the flow pattern and characteristic effects of atherosclerosis on the aorta. The model is built for very similar purposes to our model and was highly effective at simulating aortic flow in tests.

[15] Ken Kiger, "Introduction of Particle Image Velocimetry," *Burgers Program for Fluid Dynamics*, College Park: Maryland: University of Maryland (n.d.). Adapted slide deck from J. Westerweel and C. Poelma, *Technical University of Delft* (n.d.).

[16] Joseph Guzman, Lisa Novoson, Koren Roach, Matthew Rosi, "Design of a Mechanical Model for Pulsatile Aortic Flow," *Worcester Polytechnic Institute* (April 26, 2007).

1.2.2.3 Computer Models

There are many analytic physics and fluid-dynamics software that can model fluid flow in blood vessels. Some of these programs include AudoDesk Computational Flow Dynamics, OpenFOAM, and COMSOL Multiphysics. However, it becomes extraordinarily complex to model traumatic injury breakages given the wide variety of ways a vessel can be broken. Also, there are extensive variables to take into account: the non-Newtonian nature of blood, viscoelastic properties of blood vessel endothelial and connective tissue, pulsatile blood flow, blood pressure, venous vs arterial flow, vessel diameter, and many more. Mathematicians have developed multi order differential equations to model an intact cardiovascular system, but we were unable to find one that can model an injury to that system.

1.3 Statement of Project Objectives

1.3.1 Project Objective

To develop a model of the cardiovascular system that can be used as a testing platform for new traumatic hemorrhage treatment solutions. This device or methods will be cost effective and will be capable of modeling both internal and external/arterial and venous hemorrhage modalities, with the goal of practically replicating bleeding injuries in all forms accurately. In order to create a device that is as effective as possible all of the worst possible conditions of hemorrhage must be replicable. Figure 1.3 shows how our system will integrate with solutions testing.

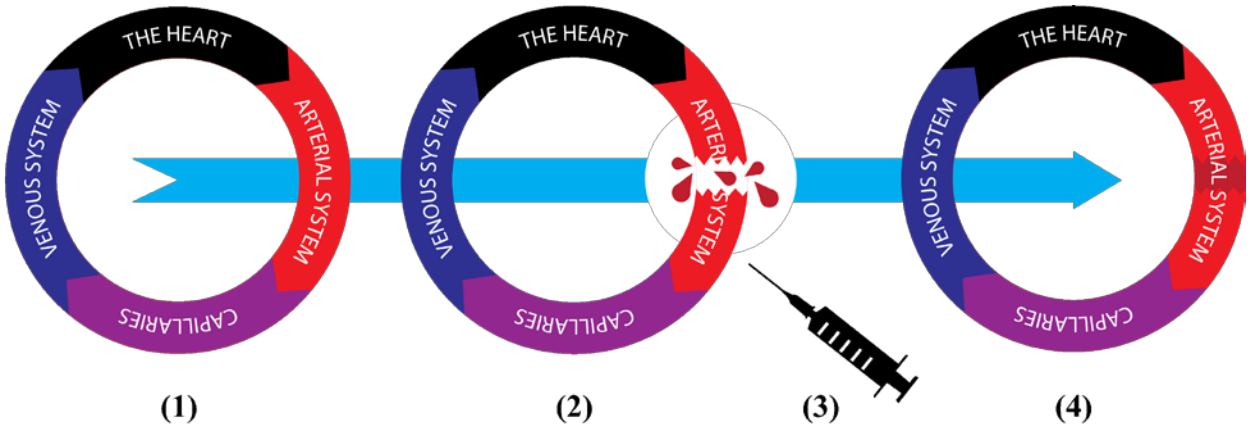


Figure 1.3 Overall Problem-solution Workflow of our Project
(1) The resting and ordinary state of the human circulatory system. Blue represents the venous system, black represents the heart, red represents the arterial system, and pruple represents the capillary exchange beds (2) The same system post arterial injury (3) Moment of administration of our proposed solution (4) The healing circulatory system after the administration of our solution candidate with restored blood flow.

1.3.1.1 Internal vs External hemorrhage

The type of bleeding must be considered. As mentioned earlier, the main focus of our project is on controlling both internal and external bleeding. Each type of hemorrhage route presents its own obstacles and difficulties in attempting to replicate their specific conditions with a cardiovascular model. Internal bleeding presents the obstacle of bleeding within the body or a closed system which means in order to really replicate internal bleeding access to the cardiovascular model must also be hindered. External bleeding, although it seems straightforward, also present the challenge of exposure to outside pathogens and open air. Exposure to outside pathogens can easily lead to infection and wounds heal faster when kept in a moist environment without exposure to air.[17]

1.3.1.2 Pressure: High vs Low

The pressure of the blood flow within the body depends completely on the area of the body at which the blood is flowing. The blood pressure of each type of blood vessel varies. Although our project is mainly focused on testing and staunching the high pressure of arterial hemorrhage, we would like for our technology to also eventually be applicable to other areas of the body where the blood pressure would be lower. If our technology can withstand high pressure then we would also expect it to be able to stanch the hemorrhage of a lower pressure. The average baseline blood pressure of arteries on the systolic phase of the heart pumping is 120 mmHg, however when a person is suffering from hemorrhage due to traumatic injury the blood pressure of the arteries can increase a great amount because of the stress the situation can cause. The systolic blood pressure of a person in a high stress situation can exceed 180 mmHg and is considered a hypertensive crisis.[18] This is a 33% increase in pressure from an average person's baseline blood pressure and calls for emergency care.[19] Our technology must be able to staunch the hemorrhage and be able to stand up and hold despite the increase in pressure.

1.3.2 Safety Concerns

Above all, safety remains a necessary concern in all aspects of our project, including the design, construction, testing, and application of the technology. The safety of our project team is vital during the design, construction, and testing of the product. The proper instructions and precautions have and will be taken to ensure the safety of our team. We have undergone extensive lab safety training to make sure we are aware of the possible hazards in the lab, in addition to learning the steps we should follow if anything were to go awry. In addition to training, there have been certain restrictions placed upon us regarding the materials we can use, for example blood. Blood is considered a biological hazard and approval from the administration and proper safety organizations is needed to work with it.[20] Since we have neither, we are unable

[17] Junker, Johan P.E. et al. “Clinical Impact Upon Wound Healing and Inflammation in Moist, Wet, and Dry Environments.” *Advances in Wound Care 2*.7 (2013): 348–356. PMC. Web. 14 Jan. 2016.
[18] "High Blood Pressure." Heart.org. American Heart Association, n.d. Web. 14 Jan. 2016.
[19] "Understanding Blood Pressure Readings." Heart.org. American Heart Association, n.d. Web. 14 Jan. 2016.
[20] "Home." *Home*. Take One Step Wellness at Work, n.d. Web. 15 Jan. 2016.

to work with real human blood and are working on creating a blood substitute to replicate the blood physical properties in the body.

Safety while applying the device, both the safety of the patient and first responder, are also extremely important. Our design and construction must be wary of the possible implications of the device. There can be no biological factors present in the solution material that can possibly interact negatively in other areas of the body. The solution must not only be effective but also must be biocompatible. Any material injected or placed in the human body from our device must either have no permanent negative effects on the body and must either biodegrade or remain inert until removed.

1.3.3 Cardiovascular Model

The cardiovascular system is very complex and difficult to replicate perfectly. There are many factors involved in the ordinary function of the heart, but only a few factors are key to testing our possible solutions. The main factors can be simplified to type of flow, pressure of the system, and vessel types. We have taken those main factors and designed a pump in which our solution can be tested on. There is the pulsatile pump replicating the flow of the heart with various types of tubing representing the different types of blood vessels. In addition to the pump and pipes of the system, the blood substitute is also extremely important.

1.3.3.1 Replication of Cardiovascular conditions

In order to replicate the conditions found within the cardiovascular system for later tests, we were required to design and build a mechanical model of the system. The system, like the actual cardiovascular system, consists of three parts: a pulsatile pump, a network of branched tubes, and a sheer thinning viscous fluid. A diagram of the system in question is presented in Figure 1.4 below.

The pump, in order to produce pulsatile high pressure flow without being prohibitively expensive, is built by hand.[21] Attached to this pump there is a network of silicone and latex based tubing that branches to simulate the bifurcation of major blood vessels in the body. Eventually, these vessels all find their way into a bucket which serves to simulate the venous reservoirs of the body (fill level of the bucket can be adjusted similarly to the volume of the venous system). A single feed line runs from the bucket back into the pump to simulate the great veins. Through this system is pumped a blood substitute composed primarily of glycogen, xanthan gum, water, and red food dye. This fluid essentially simulates the flow characteristics of actual blood without the biological hazards or financial expenses.

The system is designed to be “injured” in key ways. That is to say there are distinct and easily replaceable sections of latex and silicone tubing that can be slit or ruptured to simulate bleeding conditions. These are placed at intervals within the network to simulate various kinds and

[21] Jack Ruby, “Inexpensive, Easy to Build, Peristaltic Pump,” *Instructables.com* (2013). Web. <http://www.instructables.com/id/Inexpensive-easy-to-build-peristaltic-pump/>

severities of arterial and venous bleeding. These sections can also be placed within or under tissue simulating materials to simulate internal bleeding as the need arises.

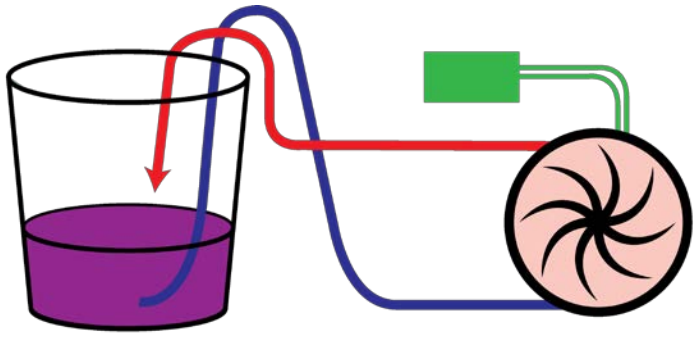


Figure 1.4 Cardiovascular Model
An overview of our proposed system inclusive of the heart simulating peristaltic pump (black on right), electronic control systems (green), venous tubing system (blue), arterial tubing system (red), and venous/capillary reservoir with blood substitute (purple)

Cardiovascular Model

2.1 Model Overview

Our primary project objective was to develop a non-biological, functional replica of the human cardiovascular system that could be used to test future bleeding trauma solutions. There were a number of secondary objectives under this primary goal that helped to further refine and define what we hoped to accomplish. Our secondary goals for this project were:

- To as accurately as possible replicate the three components of the circulatory system
- To design a model that could be adjusted and modified easily
- To design a model that could simulate injuries to itself without become obsolete
- To design a model that was simple to operate
- To keep the cost of the system as low as practicable

With these goals in mind we began to research and develop our system. To minimize wasted time and maximize our ability to progress through the design process we carried out parallel, but separate, design processes for the pump, the pipes, and the fluid of our model. Ultimately we were able to produce a device that effectively simulated the circulatory system and had the following components:

I. A pulsatile pump (simulating the heart)
II. A synthetic vasculature network (simulating the arterial system) and a synthetic collection vessel, reservoir, and fluid return (simulating the venous system)
III. A viscous blood-mimicking fluid (simulating the blood)
IV. A network of control systems and measurement devices

2.2 Component I: the Pump

2.2.1 Pump Design Introduction

The human heart is one of the most miraculous things that has ever been identified and analyzed by medical science. Replicating it for our purposes proved extremely difficult. In order to

simulate the human heart, we first looked into a variety of cardiovascular simulating pump options. Despite extensive research, we found no pumps available on the market that exactly reproduced the parameters we were hoping for. Instead of compromising on the parameters we sought, we decided to design our own pump. We elected to develop a low-cost peristaltic pump that was capable of reproducing the parameters we selected as most important.

2.2.1.1 Properties of the Human Heart

The heart's function is obviously to deliver blood to the entire human body regularly and efficiently. It does this by completely recirculating all of the body's blood – some 5-6 liters – every minute.[22] This flow is characterized as pulsatile because the flow moves out from heart in organized pressure waves or pulses. This is why you can feel the distinct cardiac beats if you hold your hand to your wrist to take a radial pulse.

Each pressure wave the heart sends out occurs when the leftmost of four heart chambers (the left ventricle) contracts and sends a small volume of blood shooting into the aorta. This volume is limited by the size of the left ventricle itself to somewhere between 60 and 80 ml. At a normal human pulse rate of between 60 and 80, this checks out mathematically. One final parameter of interest is the pressure at which blood leaves the heart. This parameter we are likely all familiar with – blood pressure, typically measured at the brachial artery in the arm, is expected to reach a maximum of between 120-140 mmHg (2.7 Psi).

From these properties we elected to focus on the flow rate of the heart and its pulsatile nature. Since the primary objective of our project is to develop a system that can simulate normal bleeding, the amount, rate, and flow pattern of blood flowing through the vessels was of a more singular importance than the pressure it was flowing at. Pressure, it was decided, could always be artificially increased later by increasing the vascular resistance of the blood vessel model instead of by modifying the pump itself.

2.2.1.2 Pump Type Selection

One of the only ways to recreate pulsatile flow without an expensive cardiovascular simulator involves using a peristaltic pump. Peristaltic pumps operate in a very unique manner – they force fluid through pipes without ever touching the fluid directly and without requiring a disruption in the flow pattern. This is ably demonstrated by the graphics in Figure 2.1 below.

These pumps are constructed by creating a circular bend in a piece of elastic tubing, using this bend to line a solid, ring-shaped housing, and then spinning a collection of symmetrically arranged rollers within this housing and against the tubing. The rollers press down on and occlude the pipe; the leading roller creates a vacuum behind it and the trailing roller creates a force on the fluid in front of it. This ultimately results in fluid being forced through the pump in rapid spurts.

[22] "Introduction to Cardiac Output," *Physiology and Pyschology*, Montana State University at Bozeman, 1998. <https://btc.montana.edu/olympics/physiology/pb01.html>

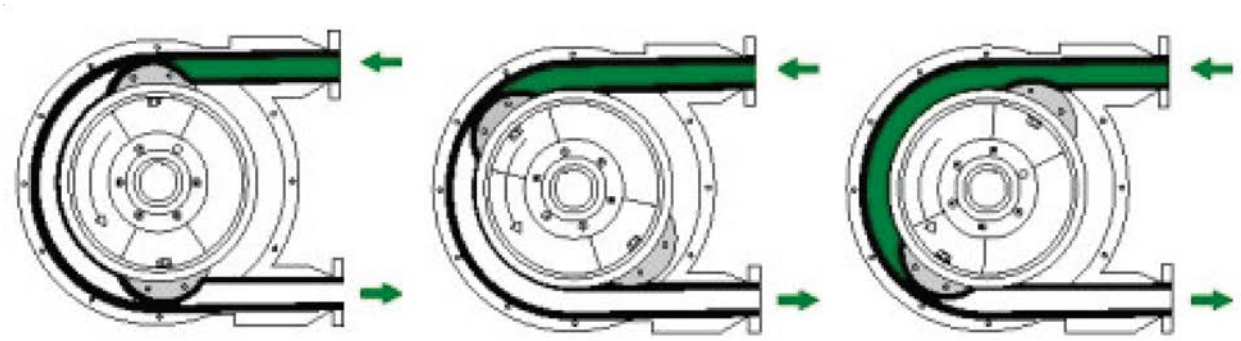


Figure 2.1 Peristaltic Pump Function
This image provides a decent description of how a peristaltic pump works. Fluid is drawn up in the inflow hose or tube by the vacuum created when the roller moves and occludes the pipe before it. Then it is pushed from behind as another roller occludes the pipe behind it. Finally, after the first roller passes the outflow point any high pressure fluid that was previously contained therein is released through the opening. Image courtesy of VMC Pumps.

2.2.2 Iterative Design Process

2.2.2.1 Design Inspiration

After extensive research, our team identified a homemade peristaltic pump model that was capable of reproducing the parameters we identified above. The pump had been designed by an online Instructable writer named Jack Ruby to move wort through the home-brewing process.[23] It was hardly a precursor to medical research use, but it had the benefit of being cheap and easy to produce, and there was video evidence documenting that it could move viscous liquid at a rate of around 5 lpm. Jack Ruby's pump is shown alongside another online model below in Figure 2.2.

Figure 2.2 Low Cost Home-built Peristaltic Pumps
The pump on the left is designed by Jack Ruby and is the design we ultimately chose as our jumping off point. The pump on the right is a similarly designed pump invented by another online homebrewer named Kyle (no last name given).[24] Both pumps work on the same peristaltic principles and use largely the same materials – only the central manifold design is different.

[23] Jack Ruby.
[24] Kyle "TheFlyingBeer," "Cake pan + skateboard wheel + bike brakes = homemade peristaltic pump??," *Homebrewtalk.com,* November 11, 2011. <http://www.homebrewtalk.com/showthread.php?t=279120>

2.2.2.2 Development of the First Prototype

Our first design was constructed almost exactly from the online design we identified. A sketch of it is shown below in Figure 2.3. The housing was constructed from a 9” diameter spring-form cake pan. In order to allow for the installation of a central axel, a 0.25” hole was drilled in the flat bottom of the pan. Two rectangular holes about 1.5”wide and 1” tall were cut into the sidewall of the cake pan 90˚ apart and centered halfway up the wall. These holes were to allow the fluid hose entry and exit points.

The next component of the pump was the central axel, hub, and roller assembly. This was constructed first by cutting standard 2 x (1.5” thick) lumber into a square approximately 3.875” on a side. This wood block functioned as the central hub for our rotary assembly. As with the bottom of the cake pan, a 0.25” diameter hole was drilled through the center of the wood block. Into both ends of this hole a 0/25” diameter size 20 t-nut was hammered. Through this now threaded hole was placed a 4” size 20 hex bolt. This would allow for the pump to be driven by a drill-powered hex driver. Two 2” casters were mounted on opposite sides of this wooden block.

This assembly was placed into the cake pan with the bolt leftover being slotted through the several washers and a hole in the bottom of the cake pan. The assembly was then carefully secured with a washer bushing and a makeshift locknut (2 size 20 0.25” nuts tightened against each other in opposing ways) in a manner that allowed rotation of the assembly but kept it in place. To complete the pump, an approximately 20” length of 0.5” diameter silicone tubing was threaded into the cake pan housing around in a loop and then out of the housing again. When the spring-form was tightened, the rollers pressed down on the pipe as desired.

The primary way that our design differed from Jack Ruby’s was that our design only used 2 casters where his had called for 4 – 1 on each side of the wood block. We believed that making this change would allow for a larger “stroke volume” (the volume of each fluid pulse produced by the pump – equivalent to the volume of tubing between two rollers) while not compromising the function of the pump. The design and change initially appeared sound, but could not successfully pump fluid.

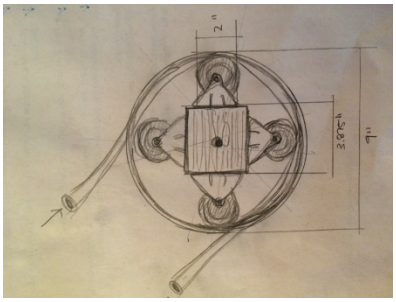


Figure 2.3 Early Pump Design Sketches
These diagrams represent some of our early designs for our pump. The image on the left is a sketch of our very first prototype which is almost entirely built based on Jack Ruby’s design.

2.2.2.3 Development of the First Functional Design

There were several possible explanations for the failure of our initial prototypes. The first possible explanation was a failure of the rollers to occlude the silicone tubing enough to create the necessary vacuum and force to pull fluid through the pump. The other possible explanation was that our 2 roller design did not provide continuous enough pressure on the tubing to maintain the vacuum and continue the flow.

In order to deal with these two possible problems, we implemented several design changes. First, we cut a new central hub out of wood. The first block was cut somewhat imprecisely; we believed that the imbalances in its shape caused tubing occlusion problems on rotation. We also added an additional two rollers to the new central hub to help keep pressure on the tubing. Additionally, we added thin layers of tape to each roller to increase their diameters and aid in their compression of the tubing. Collectively, these changes worked well. The pump functioned, albeit inconsistently, when driven by a corded drill. It is shown directly below in Figure 2.4.

Figure 2.4 The First Functional Pump Prototype
Note that the design has not changed significantly in appearance from the design sketches in Figure 2.2. Note that there is duct tape layered around each of the wheel rollers to give them additional width and bite on the tubing. This is the result of measuring errors and equipment (skill saw) imprecision accumulated during construction.

2.2.2.4 Development of the Final Design

Our final design implemented still more changes. In order to dress up and render the internal machinery more precise, we had a central hub precision-milled out of High Density Polyethylene (HDPE). We also milled a special drive axel out of recycled steel that interfaced with the HDPE block on one end and a set screw connected to an electronic motor on the other. We also built a larger wooden housing to hold our spring-form pan housing and the electric motor that would drive our pump. The electric motor was removed from a battery operated drill. This design is easier to communicate graphically, as is accomplished in Figure 2.5 following this paragraph.

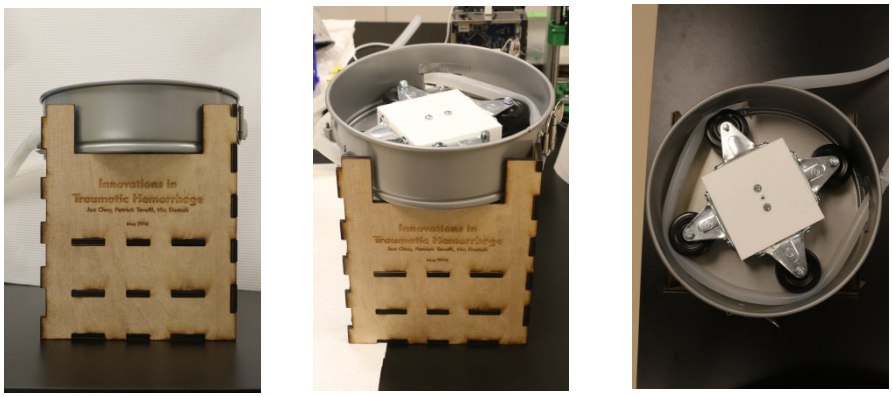


Figure 2.5 External Shots of the Final Design and Housing
The picture on the far left shows the front of the housing and final pump. Not much can be seen in this image, but the more oblique top/side view in the middle begins to show the pump head and piping. The final image on the right shows a top down view of the pump head.

2.2.3 Final Design Description

The final pump design is similar to the original models in appearance, but significantly more improved. The pump housing is still constructed from a 9” diameter spring form cake pan. The housing shell is broken in two places, approximately 90 degrees (~7”) apart, by rectangular cutouts that fall exactly halfway up the shell and are sized to be around 2” wide and 1” tall. The housing base (bottom of the pan) is broken in its exact center by a 1.5” diameter hole. This is meant to allow the passage of an adapter that connects the drive system to the pump head.

The pump head itself is constructed by mounting 4 independent 2” caster wheels to the 4 small faces of a cube of white HDPE plastic milled to be exactly 3.875” x 3.875” x 1.5.” Three 0.125” holes are drilled through this block in a line from one large face to the other (perpendicular to the casters). Each hole is 0.5” from the next and the trio is centered with the middle hole drilled through the exact middle of the block. Two 2” long machine screws are eventually passed through the outside holes to secure the block to the drive adaptor with a third screw, placed through the middle hole, securing the drive adaptor to the drive shaft exiting the motor.

The drive assembly begins with the removal of an electric motor from a Black and Decker 12V cordless drill. Our drill motor had a forward protruding drive shaft that was threaded – most will probably have the same front adaptations. An 1.5” diameter cylindrical adaptor milled out of solid steel was then threaded partway onto the drive shaft of our motor. The adaptor was then secured here with a smaller screw which, as described earlier, passed all the way through the plastic block, partway through the steel adaptor, and into a small tap in the center of the end of the driveshaft. This adaptor was then further secured to the block with the aforementioned screws passed through the two outside holes of the plastic from the upper face.

This motor assembly was then secured to the two horizontal panels of our pump support box and housing. Two custom printed brackets were placed around the motor in such a way as to inhibit any vertical or horizontal movement of the motor itself relative to the two wood panels. The

remainder of the box was constructed by box jointing 4 side panels and a floor panel to the two interior support panels. All of these panels were secured except for one of the side panels which was left unsecured to ensure easy access to the drive system and the interior of the pump support.

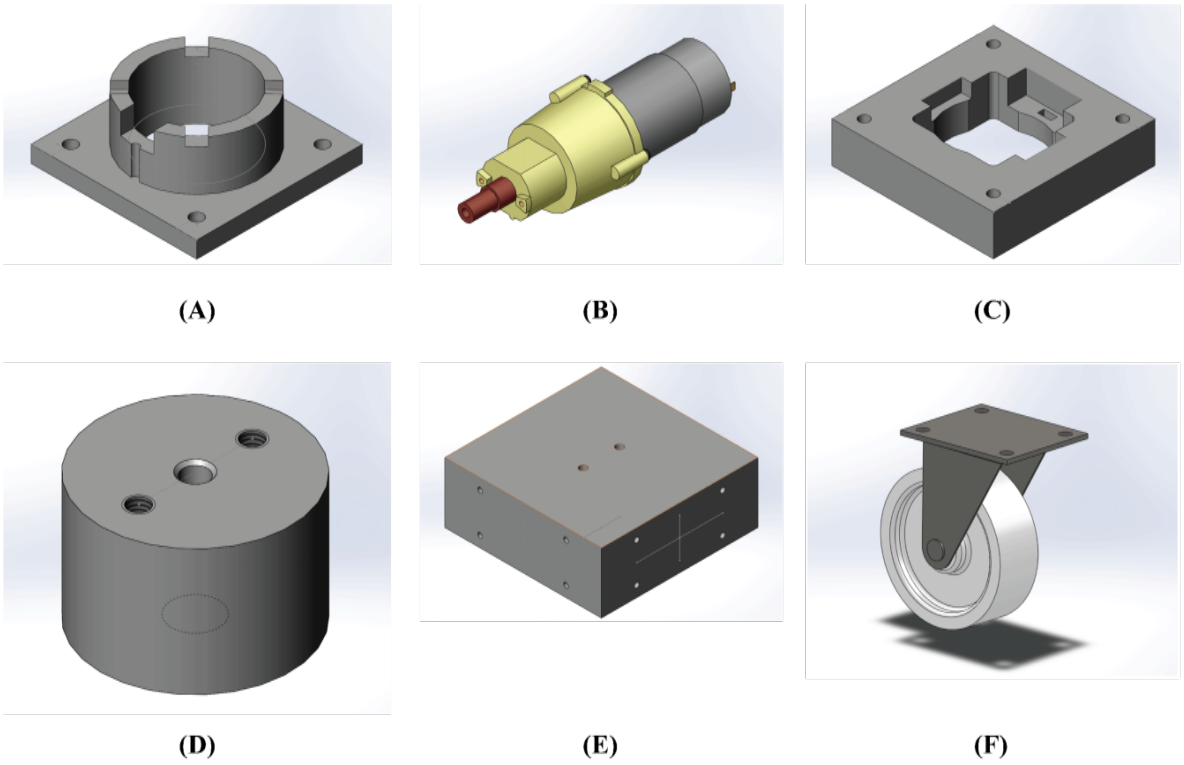


Figure 2.6 Final Pump and Housing Part Diagrams
(A) Rear mounting bracket for drill motor. The four corner holes allow the bracket to be secured into a horizontal panel of the housing below the motor preventing it from sliding backward into the panel. (B) A Solidworks model of the drill motor itself. (C) Front mounting bracket for the drill motor. The four corner holes allow this bracket to be secured to prevent the drill from moving upward into the pump itself. (D) The adaptor that joins the drill drive shaft to the pump head central block. (E) The pump head central block that is milled to accommodate connection to the adaptor and four casters. (F) A Solidworks model of the Waxie 2" rubber casters used as rollers in our pump.

2.3 Component II: the Pipes

2.3.1 Pipe Design Introduction

The human body is possessed of a unique and complex network of blood vessel responsible for delivering blood to its various tissues and organs. Almost all injuries that present with hemorrhage involve some injury to this vasculature network, and so it was incredibly important that our device include a vasculature simulating network which could be easily injured to simulate such wounds. That being said, no human attempt to exactly replicate the cardiovascular

network could succeed – there are over 60,000 miles of blood vessels within the average adult.[25] Instead, we chose to replicate those properties crucial to the simulation of very dangerous bleeding injuries. To do this required examining the properties of the vasculature. Ultimately we decided on a very simple model that sought only to replicate the arterial branched network to the level of small arteries.

2.3.1.1 Properties of the Human Circulatory System

The network of blood vessels in the human body consists of two loops of vessels. The pulmonary loop distributes blood to the lungs, and the systemic loop distributes blood to everything else. Since the vast majority of bleeding injuries occur in the systemic loop, and since the pulmonary loop merely replicates the systemic loop on a smaller scale, we chose to recreate only the systemic loop in our model.

Each loop of the vasculature is further characterized by the direction its vessels take the blood. Arteries direct highly oxygenated blood away from the heart and into the tissues, while veins remove deoxygenated blood from the tissues and carry it back to the heart. The two types of vessels are joined by capillaries where gas exchange takes place.

Arteries carry blood at high pressure – the pressure can be over 120 mmHg (2.3 Psi).[26] The veins carry blood at much lower pressure. This is very important when one considers bleeding. Venous bleeding injuries tend to ooze, or bleed slowly, because the blood is at such low pressure. By contrast, arterial injuries typically bleed furiously and quickly. Individuals suffering from this type of bleed can lose a fatal amount of blood in minutes, and this is the type of bleeding we chose to focus on as a result. Any bleeding treatment that functions in the high pressure environment of the arterial system will be logically functional in the venous system as well.

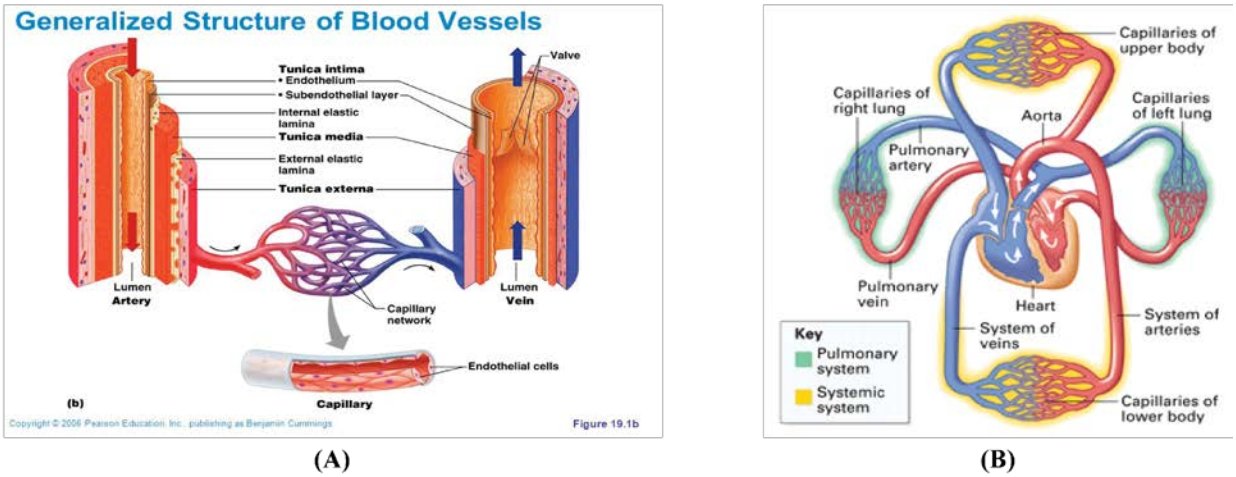
The venous system serves as more than just a fluid return system in the body – it also serves as a fluid storage center. The veins can dilate or constrict in order to change the total volume of the circulatory system. This function is important to our model, as the amount of fluid circulating through the system will be changing as it would in a real bleeding situation.

Having a system that can adjust its volume helps maintain the integrity of that system as it experiences volumetric shifts. Luckily, our system can accomplish this using nothing more than a bucket and a single hose. There is no need to simulate the entire venous network – the single pump inlet pipe drawing from a bucket whose volume is allowed to fluctuate suffices.

Table 2.1 Human Arterial System Blood Vessel Characteristics[27]

Vessel	D (mm)	D (in)	Wall thick (mm)	Wall thick (in)	L (cm)	L (in)
Aorta	12.5	0.49	2.00	0.08	50	19.7
Arteries	2.00	0.08	1.00	0.04	50	19.7
Arterioles	0.01	0.0004	0.02	0.0008	1	0.39

[25] B. W. Zweifach, "The Microcirculation of the Blood," *Scientific American* 200, 1, 1959.
[26] Mistovich and Karen.
[27] V. Shanthoshini Deviha, P. Rengarajan, R. Jahir Hussain, "Modeling Blood Flow in the Blood Vessels of the Cardiovascular System Using Fractals," *Applied Mathematical Sciences* 7, 11, 2013. 527-537.



(A) (B)

Figure 2.7 Human Circulatory System Anatomy
(A) This image shows the anatomy of the blood vessels in the body. On the left you can see a cutaway of an artery and on the right you can see the cutaway of a large vein. These images do not show the blood vessel wall thicknesses to scale. (B) This image shows to two main loops of the cardiovascular system. Vessels highlighted in yellow are part of the systemic circuit while those backed in gray are part of the pulmonary system.

2.3.2 Selection of Vascular Material

There were a huge variety of materials available to us that could be used to simulate blood vessels in the human body, but we were primarily interested in materials that demonstrated the following properties:

- Elastic
- Durable
- Biologically inert
- Inexpensive
- Easy to work with

The latter two made good sense and were non-negotiable; this was supposed to be a low cost device to lower the barriers to entry to innovation in the field of traumatic hemorrhage. It would not have been any good for the model to use a very expensive, but realistic, vasculature simulating material. It was also supposed to be easy to put together and repair, which meant that it couldn't use materials that were difficult to get a hold of.

The others are perhaps less intuitive, but equally as important. The material had to be elastic because the fake blood vessels had to be similar in nature to the real ones and real vessels are elastic by nature. Any tests performed on this model in the future could have been invalidated if the blood vessels did not accurately enough simulate blood vessels. The material had to be durable so that it could withstand extended use and abuse outside of the protection of a cleanroom or sterile laboratory. This model will be used to test solutions that will be required to function in the elements (in the field) so the model itself must function in the field as well. Finally, the material selected had to be bio-inert so that future biological tests and solutions could be used or run in coordination with this model.

A project similar to ours that we mentioned earlier in the introduction to our thesis used natural rubber "Penrose" tubing to simulate the arteries and blood vessels in their design as they found the elastic and material properties of this kind of tubing to be most similar to that of the blood vessels.[28] Penrose tubing is used extensively in medicine and can be used in highly invasive surgical procedures safely. Unfortunately, it is very hard to get in reasonable lengths and quantities. It is generally sold pre-cut so that it can serve as a drain during surgical procedures. As such pre-cut, collapsed tubing was not well-suited to our purposes we decided against using it in our project for the current phase, but it is being held up as an option for future research.

These criteria limited the available pool of materials significantly, and we decided to further limit the pool by looking only at tubing varieties currently in use in medicine and biological research. There were only a few options that qualified under that change in criteria. These options are listed in Table 2.1 along with a number of their material properties. Some of them are also shown in Figure 2.8 on the next page.

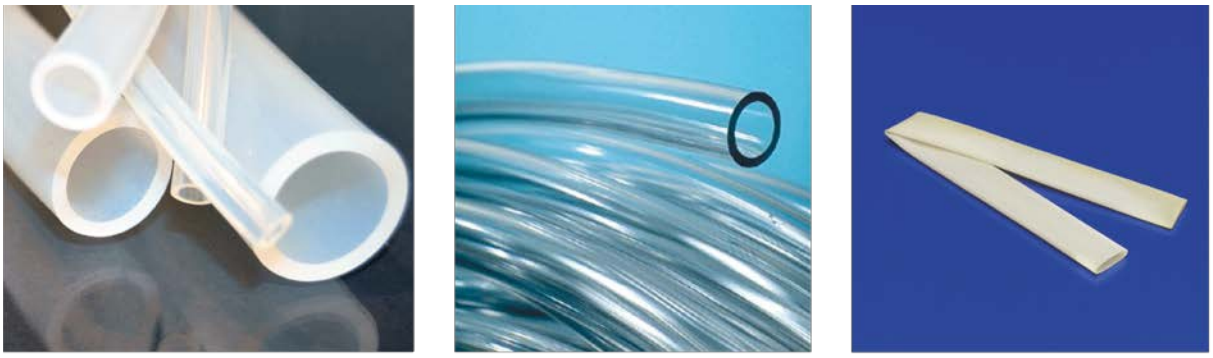
Table 2.2 Comparison of Medically Accepted Rubbers[29]
Please note that Weather, Wear, and Elasticity are all rated on a 1-5 scale. Cost is rated on a $-$$$$ scale. The elasticity category is rated based on the material's demonstrated ability to rebound to its original shape and under elongation and should not be confused with its Young's Modulus which is a measure of how much the material deforms under a constant stress load. This information is taken from several sources – the scoring is subjective based on the data collected.

Material	Cost	Weather	Wear	Elasticity	Young's Modulus[30]
Silicone	$$$	5	2	5	0.005 - 0.02
Natural Rubber	$$	2	4	3	0.0015 - 0.0025
PVC/Vinyl	$$	5	4	3	2.14 - 4.14
Nitrile[31]	$	2	4	4	0.002 - 0.005

From the information we collected, we determined that the best fit was silicone tubing. Although silicone scored low in the "wear" category, it scored significantly higher than the other materials in the elasticity and weather durability categories. Ultimately we deemed that it was worth the slightly greater expense and risk of damage to use a material with better elastic properties and better weather resistance. Silicone is also considered a more appropriate medical material because of its high biocompatibility.

Some of the other materials are not as well tested in vivo and are less acceptable for highly invasive medical use. Nevertheless, in the testing phase of our project we did test the material properties of silicone tubing against those of vinyl (PVC, a very medically accepted material and a very easy to get a hold one at that) to ensure that we made the correct choice.

[28] Guzman, et al. "Design of a Mechanical Model for Pulsatile Aortic Flow"
[29] "Rubber Properties," Mykin Incorporated, 2016. < http://mykin.com/rubber-properties>
[30] "Materials Data Book," Cambridge University Engineering Department, 2003
[31] "Nitrile Rubber," Matbase, 2004. < http://www.matbase.com/material-categories/natural-and-synthetic-polymers/elastomers/material-properties-of-nitrile-rubber-nitrile-butadiene-rubber-nbr.html#properties>



(A) (B) (C)

Figure 2.8 Different Medical Tubing Options
(A) Silicone tubing in a variety of sizes. Silicone is often identifiable because it has a very distinctive semi-translucent coloration and rigid wall structure. (B) PVC or Vinyl tubing in a standard diameter. PVC is distinctively transparent in its natural form, but can be colored. (C) Penrose tubing, easily identifiable by its lack of rigidity and natural rubber coloration.

2.3.3 Iterative Design Process

2.3.3.1 Development of Initial Vasculature Model

Our initial vasculature model was composed primarily so that we could determine the function of our pump and plan future upgrades for the system. It consisted of only one long piece of 0.5" diameter silicone tubing that was passed through the pump and ran from one of two buckets to the other. This model semi-accurately recreates the blood flow conditions present in large arteries like the Brachiocephalic Trunk, the Common Iliac, and the Carotid. Since most dangerous and serious bleeding injuries have to do with at least one larger artery, this was taken as an acceptable model for initial tests.

2.3.3.2 Addition of Measurement Components
For our next prototype we decided that we needed to add in some methods of collecting data based on our flow. We purchased a small threaded inline flow meter for our device that could talk to our Arduino Uno and ultimately modify the system based on readings from flow data (speed up/slow down the drill motor drive). This inline flow meter was coded to record data in liters per minute to aid in the diagnostics and make it easy to determine if we were hitting our particular benchmarks (pulsatile flow at 5-6 lpm speed).-

We also wanted to attempt to do pressure analysis and system response. To this end we purchased a small selection of force based pressure sensors that we thought we could add into a dead end branch of the system in order to collect pressure data. This did not prove as easy as we expected, but we were able to install a new pressure sensing dead end. Unfortunately, it was not as easy to set up the recording process and record data already in the right units for these sensors, nor was it easy to get them to read properly, so we eventually abandoned the pressure project altogether. More information on these two components is given below.

2.3.3.3 Development of Final Design

One thing lacking in our relatively few previous iterations of vasculature model was any sort of arterial branch modeling. All of our previous models had assumed a single, isolated large artery. For a variety of reasons, this rendered our testing inadequate. Without branching, there was no way to ensure that our model behaved like the human circulatory system when injured. In the human body, an injury in artery A will influence flow patterns and blood behavior in arteries B through K. Similarly widespread effects are again possible when a solution is applied to said injury. We needed to more completely model arterial branching.

We decided that a pattern of branching based on the power of 2 would be ideal for our model. It would allow us to keep the total number of "vessels" at the most downstream end of our arterial system more manageable than other exponential models while being more accurate than a linearly increasing branched model. We purchased a number of different sizes of silicone tubing to be used in our model. Silicone tubing only comes standard (at a reasonable price) in 0.125" inner diameter increments down to a minimum inner diameter of 0.125". We purchased every available inner diameter size in between this 0.125" ID minimum and the 0.5" ID maximum our final pump model allowed.

We also purchased a quantity of polypropylene fittings for our tubing in order to allow for the more complicated vasculature model layout. We developed a complete network that branched 4 times and decreased in diameter 3 times. We also added an injurable section so that a portion of the system could be "injured" and then easily replaced without affecting the entire model. No changes were made to the initial length of 0.5" diameter tubing that passed through the pump and the inline flow meter/pressure sensor.

2.3.4 Final Design Description

Our final design used 4 different diameters of silicone tubing in a branched model to gradually divert a fluid flow of 5-6 lpm from a simulated large artery (0.5" diameter) to 16 simulated small arteries (0.125" diameter) according to the map in Figure 2.9 below.

Each node on the diagram below represents a bifurcation in the preceding tubing network that splits a single tube of a set diameter (indicated by the region that tube is in) into two tubes of the same diameter. Shortly after the tubes were divided in this manner reducer couplings were inserted to reduce the diameter of the two newly created tubing branches to the next lowest diameter level. This does not represent an ideal branching method, but it is the best that can be achieved without developing one's own polypropylene tubing fittings (there are no reducing "Y" splitters – the two goals must be achieved with separate fittings in sequence).

After being reduced by the special couplers, a predetermined length of tubing with a set diameter and no branches was attached downstream of the node to each new branch. The lengths these pieces of tubing took were determined so that the fluid flow within these sections would reach its fully developed state. In fully developed flow, the velocity profile of a moving fluid does not change over time. This was desirable in our model because it reduced the number of variables that could affect flow at an injury site and confound future testing results.

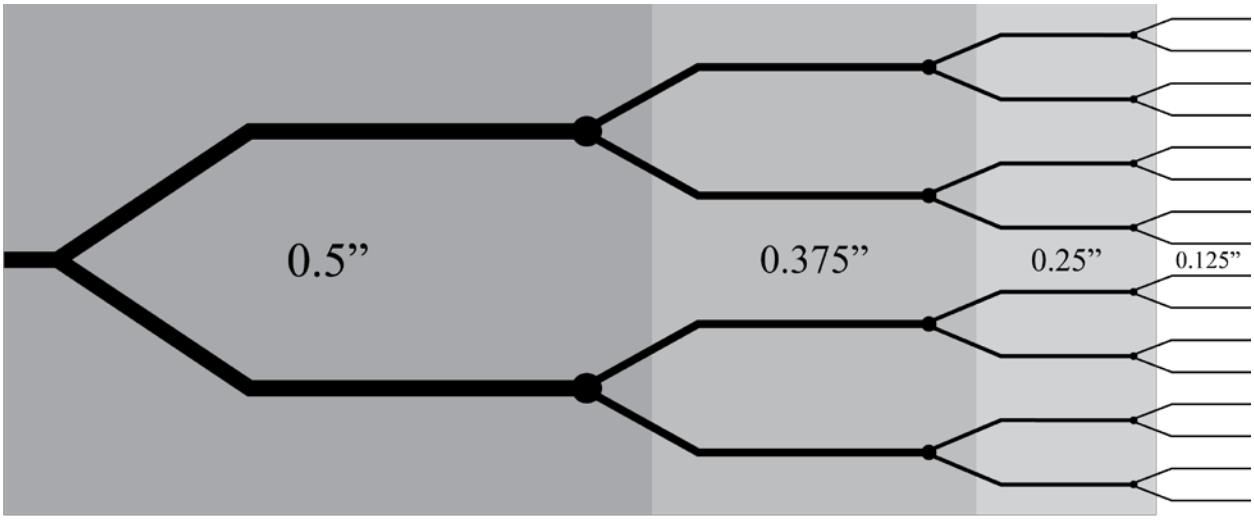


Figure 2.9 Vasculature Model Map
This is a map describing the characteristics of our vasculature model. Each number indicates the inner diameter of the silicone tubing in that colored region. Note that the colored regions actually extend beyond where the line size changes - this is accurate to real life. The tubing size change occurs via downcoupler downstream of the splitter node but this could not easily be graphically represented. Each node indicates a “Y” fluid splitter.

The appropriate length a tube or pipe must have to allow for fluid moving within to develop fully is called the entry length. The entry length can be determined using either of the following equations depending on if the fluid is in turbulent (messy) or laminar (unidirectional and tidy) flow. Note that L_e = entrance length, D = pipe diameter, and Re = Reynold’s number.

For laminar flow, $$\frac{L_e}{D} = 0.06 * Re \quad (2.1)$$

For turbulent flow $$\frac{L_e}{D} = 4.4 * \ Re^{1/6} \quad (2.2)$$

Though factually true, these equations were not ultimately of use to us because we did not, and could not, calculate Reynold’s number until we confirmed that our device could actually pump fluid at the flow rate we desired. Luckily, it is excepted and standard practice in fluid dynamics to assume that the entrance length for a given pipe is never greater than the number given by:

$$L_e \ \leq 10 * D \quad (2.3)$$

This made calculating entrance lengths vastly easier for us. The results of those calculations are summarized in Table 2.1 below. We planned on including at least one injurable section per diameter of tubing included in the model – a total of 4 different injurable sections. In order to ensure that the amount of variables uncontrolled by our model was at a minimum, we desired to achieve fully developed flow upstream of, within, and downstream of these injurable sections. For this reason we adopted tubing lengths that were slightly more than 3 times the required length (thus allowing the injurable branch to be divided into three segments, one of which would be easily replaceable, without affecting the fully developed flow parameters). This is why the selected lengths (shown also in Table 2.2) seem so much higher than they need to be.

Table 2.3 Entrance Lengths and Selected Lengths for Various Tubing Diameters
Note the dramatically longer selected lengths as discussed in the previous paragraph.

Diameter (in)	Min Le (in)	Length Selected (in)
0.500	5.000	18.000
0.375	3.750	15.000
0.250	2.500	10.000
0.125	1.250	4.000

Thus, our final design can be mapped as shown in Figure 2.6

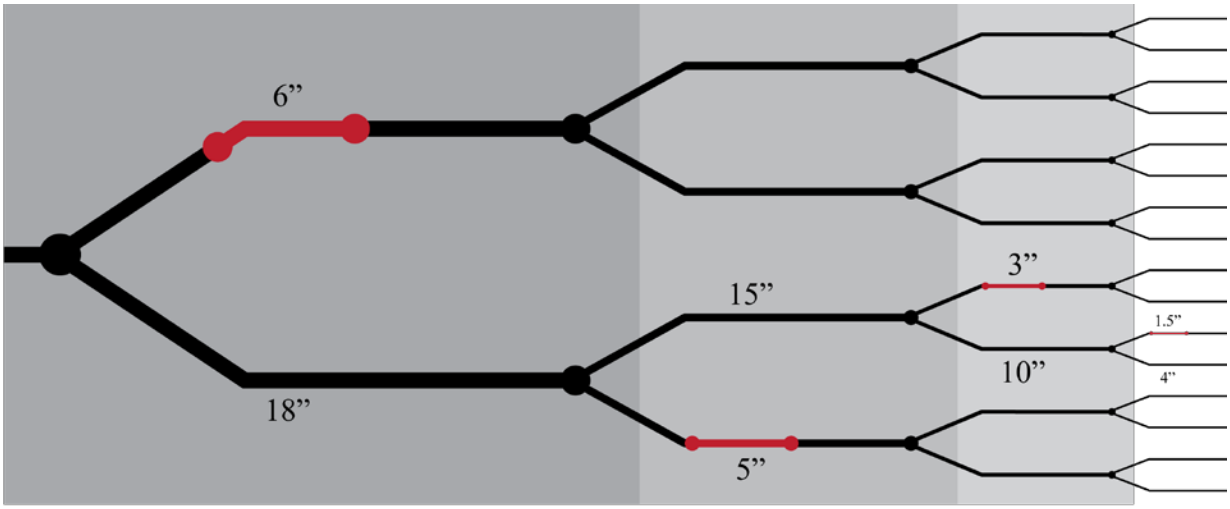
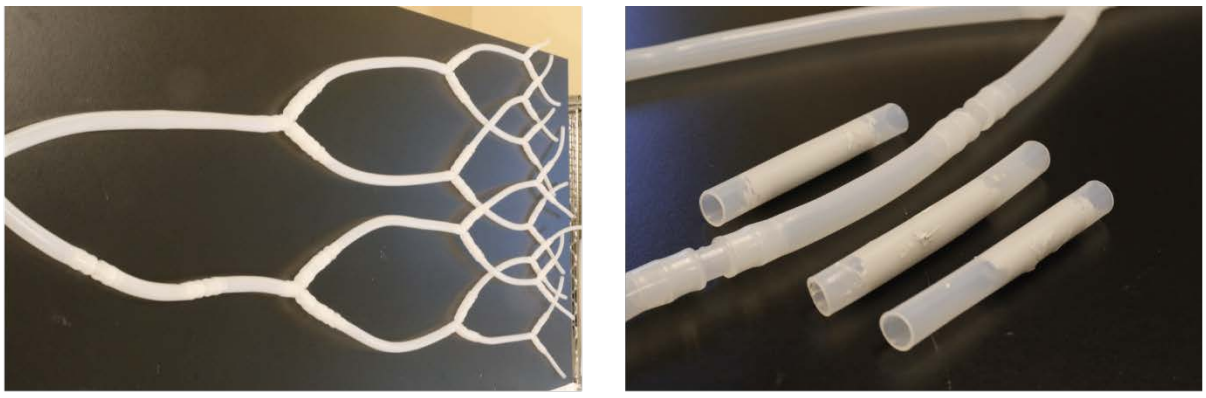


Figure 2.10 Vasculature Model Map Showing Tubing Lengths
Injurable sections are pictured in red with their own lengths. Any noninjured lengths of tubing within the same color region have the same length. Note that no two injurable sections are ever within the same branch to minimize the potential that changes caused by one might confound behavior at another. The diameters of the tubing remain the same as indicated in the previous figure.



(A) **(B)**

Figure 2.11 Vasculature Model and Injurable Sections
(A) This image shows a picture of the vasculature model in its entirety. The outlet tube from the pump is not visible, but the two 0.5” tubes which split from the first bifurcating node are clearly in frame on the left of the image. It is also just possible, if one looks at the lower 0.5“ branch, to make out the injurable section which is shown much more clearly in (B) with a number of used injurable sections (that have been injured and treated) next to it.

2.4 Component III: the Fluid

2.4.1 Fluid Design Introduction

In order to complete a functional model of the cardiovascular system, blood or a blood-mimicking fluid is necessary. Blood or a blood-mimicking fluid is necessary to provide the proper flow characteristics under pulsatile flow conditions that the cardiovascular system experiences. The characteristics of blood that we were looking for were viscosity, non-Newtonian shear thinning behavior, and pH.

While human blood is the ideal fluid because it is precisely what flows through the human cardiovascular system, blood was unnecessary and inferior to a blood-mimicking fluid for a number of reasons. The primary reason comes from the goal of this project. This objective of this project is to develop a dynamic model of the cardiovascular system to test physical, bio-inert substances that can stop traumatic bleeding. It is not necessary then for the fluid to have any biological component; in fact a biological component would unnecessarily complicate the development of the solution.

The other considerations were logistical: we would need multiple liters of blood which would be expensive beyond our budget, using a biohazardous substance like blood would require special lab facilities and slow down testing time, and we would have to have additional controls to control the temperature of the blood. With a blood mimicking fluid at 21 degrees C we were able to have the same physical properties of blood at 37 degrees C. Taking this into account, we decided to design and use our own.

2.4.2 Iterative Design Process

After the decision was made to use a blood-mimicking fluid, we followed the following iterative design progression: competing product analysis, literature review, Newtonian fluid development, non-Newtonian fluid development, and pH adjustment.

2.4.2.1 Product Analysis

Our product analysis found that there were very few substances on the market that mimic solely the physical properties of blood. There are a number of emerging products for clinical applications including solutions focused on O2 transport and bio-inert volume expanders for traumatic blood loss situations.

There was one product that interested us, a blood-mimicking model 046 created by CIRS Tissue Simulation and Phantom Technology used for velocimetry flow phantom testing.[32] However we found this blood-mimicking fluid to match the viscosity we would like, but it did not exhibit the shear-thinning properties of blood nor did it have physiological pH. Additionally, we found that we could develop our own solution for less cost and matched more of the characteristics of blood we were aiming for.

2.4.2.2 Literature Review Findings

After deciding to develop our own fluid and not purchase a product, we conducted a thorough literature search, as described in section 1.2.2.2 consisting of multiple blood substitutes used for clinical applications and flow phantoms. Most of the relevant literature focused on a non-Newtonian blood-mimicking fluid consisting of 40% Glycerol and 60% water[33], but two research papers suggested the use of xanthan gum as shear thinning agent in a water, glycerol solution.[34] Using this research as a starting point, we began to develop our fluid.

2.4.2.3 Newtonian Fluid

The first iteration of our blood-mimicking fluid design was designed to repeat the Newtonian glycerol solution we found in literature. In larger arteries and the aorta, blood exhibits Newtonian properties, so a simple glycerol-water mixture would be useful for testing these vessels, but our robust vascular model included more than just the arteries. We conducted viscosity testing to validate what we found in research. The results of this testing and the testing protocol can be found later in Section 3.4. A comparison of Newtonian and non-Newtonian fluid properties can be found in Figure 2.12 later in this section.

2.4.2.4 Non-Newtonian Fluid

A non-Newtonian shear thinning fluid is necessary for an ideal flow profile of our pulsatile model. Non-Newtonian properties affect the fluid dynamics in a few ways that are essential for a testing platform. First, it affects the sustainability of laminar flow in a pulsatile flow system, which affects how the fluid dynamics are affected by injuries to the system or blockages to the vessels, which has implications for injectable biomaterials. It also affects the entrance length until fully developed parabolic flow occurs, a parameter which was important in determining the length of our injurable sections. We accomplished this by adding xanthan gum to a solution of water and glycerol to create a shear-thinning viscous fluid. Viscosity testing and a plan for quantitatively determining the shear thinning property can be seen later in Section 3.4 and 4.3 respectively.

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[32] "Blood Mimicking Fluid Model 046," CIRS Tissue Simulation and Phantom Technology, reproduced from recipe provided by K. V. Ramnarine, D. K.Nassir, P. R. Hoskins, J. Lubbers, "Validation of a new blood-mimicking fluid for use in Doppler flow test objects," *Ultrasound in Medicine & Biology*, 24, 3, 1998. 451-9 <http://www.cirsinc.com/products/all/72/blood-mimicking-fluid/>

[33] K. A. Brookshier, and J. M. Tarbell, "A Method for Matching the Refractive Index and Kinematic Viscosity of a Blood Analog for Flow Visualization in Hydraulic Cardiovascular Models," *Biorheology*, 126, 2004. 529-35.

[34] T. T. Nguyen, Y. Biadillah, R. Mongrain, J. Brunette, J. C. Tardif, and O. F. Bertrand, "Evaluation of a Transparent Blood Analog Fluid: Aqueous Xanthan Gum/Glycerin,"*Biorheology*, 30, 1993. 107-16.

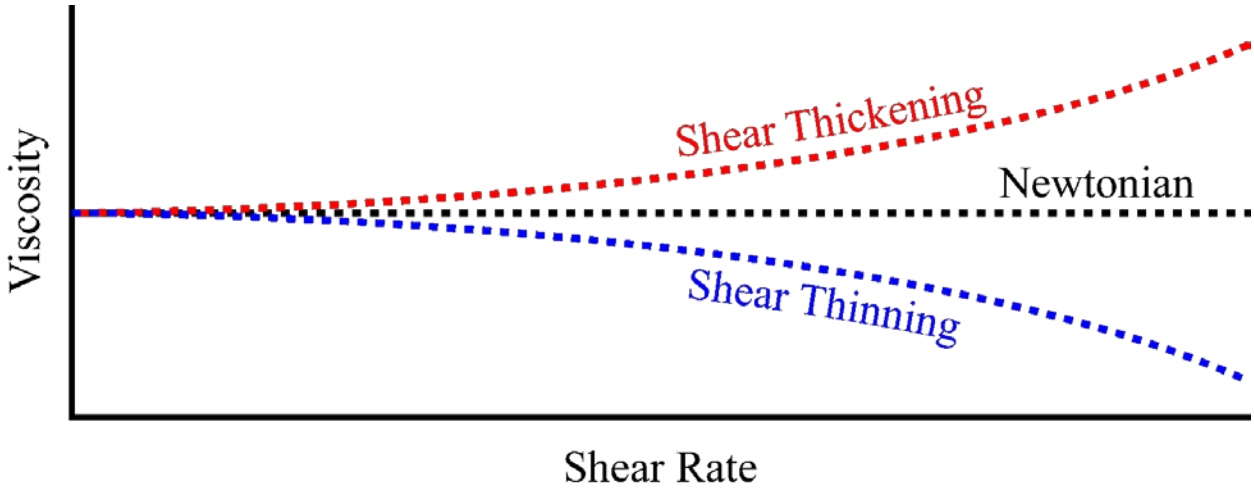


Figure 2.12 Comparison of Different Fluid Type Properties
This figure demonstrates the relationship between Newtonian and non-Newtonian fluids nicely. In practice, shear rate's effect on fluid viscosity is closely paralleled by the impact of shear force on fluid viscosity. This image was inspired by a similar one visible in the article "Viscosity of Newtonian and non-Newtonian Fluids" found at RheoSense.com

2.4.2.5 pH Adjustment

The final step in completing the parameters for our shear-thinning blood mimicking fluid is to adjust the pH from neutral to 7.4 to more accurately replicate the pH found in blood. A correct pH is important for the fluid because it can allow for the experimentation of pH sensitive hydrogels, sol-gels, and other biomaterials as potential solutions for stopping traumatic bleeding. In order to accomplish this, we replaced the water base with a phosphate buffer consisting of NaOH and KH2PO4 in aqueous solution with the same concentrations of glycerol and xanthan gum. The testing results can be found in section 3.4.2.

2.4.3 Final Design Description

Our final solution was a colloidal mixture of phosphate buffer, glycerol, and xanthan gum. This solution provided all of the physical blood characteristics we needed for our system: viscosity, shear-thinning property, and pH. The testing we conducted consisted of viscosity testing and pH testing which will be described in detail later in Chapter 3.

2.5 Integration and Control Systems

2.5.1 Component Integration

The Arduino system allow multiple applications and devices to be run at once through the same program code. Both the flow meter as well as the pressure sensor is wired through a breadboard to connect to the arduino board and give back quantitative data. In our final platform design the pump will also be controlled through the arduino program and will be able to run simultaneously. The pump as well as the data collection program for the system will be able to run in series so continuous and accurate flow is achieved. The feedback loop would begin with the flow meter readings interacting with the running code and if the reading were above or below a certain set threshold, then the program code would run through a different loop and increase or decrease the flow rate accordingly. Component Integration is the featured subject of Figure 2.13 in the next section.

2.5.2 Control Systems

2.5.2.1 Arduino System Control

The electronic components of our platform were meant to obtain quantitative data as well as serve as a control for the motor system. We chose the Arduino system to control all of this because of its extreme versatility as well as the general low cost of the Arduino system and additional parts. It is an open source programming system meaning that anyone can use the system and share the code to further the possibilities of what can be done with the system and technology. It is also readily compatible with any computer system and connected through the USB port. We chose to use a basic Arduino Uno board. The Arduino system has its own programming language that is based off of a language called Processing, but it is also compatible with and supports the programming languages of C and C++.

2.5.2.2 Feedback Loop for Motor Control

From the beginning of the design process we envisioned that the motor of our system would be completely automated. Using an adapted motor from an electric drill we were planning on attaching it to a power source that would be able to provide the necessary power, which would most likely be a transformer, and then controlling it using the Arduino board and a technique called Pulse Width Modulation. In summary, this technique would send extremely rapid signals to switch the motor on and off and depending on the intervals between those signals, the motor speed would be adjusted.

In addition to controlling the motor through PWM, we would create a feedback loop within the code. This feedback loop would read the flow rate in liters per minute and if it were above a certain threshold like 7 liters per minute or above, then the code would decrease the frequency at which the PWM was operating at, in turn slowing the motor and flow rate. If the system read the

flow rate to be below a certain threshold like 4 liters per minute and below, then the code would increase the frequency at which the PWM was operating at, in turn increasing the motor speed and the flow rate. Unfortunately, we were unable to bring this entire system together in a functional form. Our electronic control systems are still not functional, and our platform still has to be powered with a drill or other handheld power source.

2.5.3 Measuring Equipment

2.5.3.1 Flow Meter

The flow meter we selected to use in our design was a ½'' plastic NPS threaded Arduino liquid flow meter. It was able to read the flow rate once every second and calibrated to be precise within 10%. This flow meter takes advantage of the Hall effect and counts the pulse output it creates, by counting the pulse output the meter can track the amount of fluid flowing through the meter. It was easily connected to the Arduino and power source because it can operate with a working voltage of 5V which equals the max voltage output on the Arduino board. The code used was adapted from the Arduino Open Source Code library and changed to fit our purposes.

2.5.3.2 Pressure Meter

There were two possible pressure sensors that we could have used for testing. The first was a Round Force-Sensitive Resistor and the second was a larger area Square Force-Sensitive Resistor. Both sensors operate by measuring the change in resistance when pressure is applied to the sensor and it converts that resistance into a readable pressure. Given the design of our pressure sensor hub the Round Force-Sensitive Resistor was the sensible choice to integrate into the hub. The sensor was also easily connected to the Arduino Uno board through a breadboard.

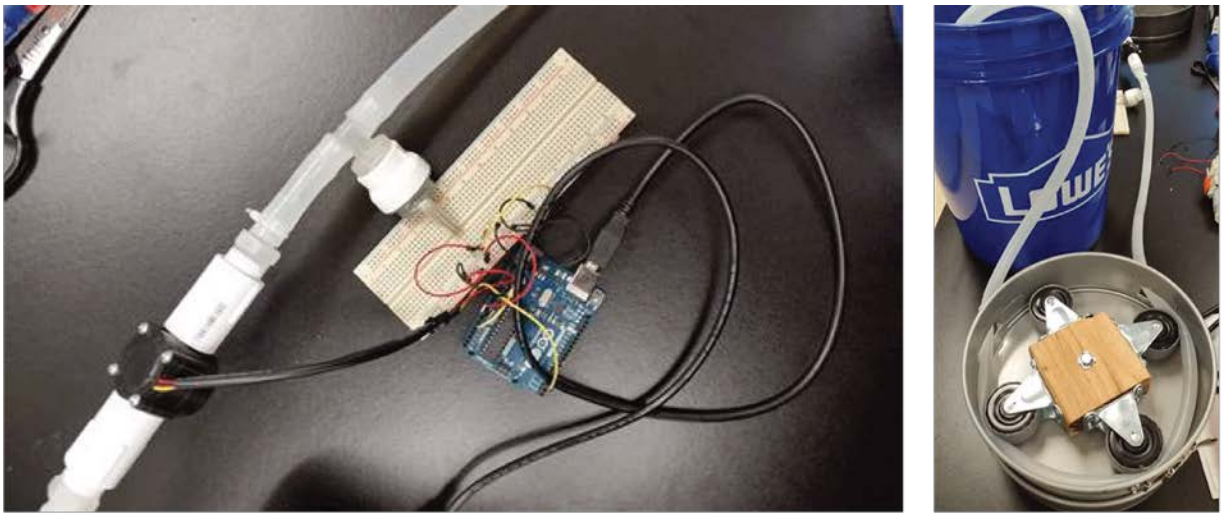


Figure 2.13 Electronic Controls, Measurement Systems, and Component Integration
The image at left shows the in-line flow meter (black cylinder in silicone tubing), the make-shift fluid pressure sensor (white cylinder coming off of pipe T-junction in middle of image), and the electronic data recording system (Blue Arduino and cream breabboard) that connect each to a computer. The image at right shows how the system is integrated. You can clearly see the pump in the foreground, the vasculature network (including the bucket) in the mid-ground, and the electronics peaking around the corner of the bucket in the background.

System Testing

3.1 Testing Introduction

Every scientific research project or engineering endeavor must involve rigorous testing to ensure its validity and direction. Our project is no exception. After conceptualizing our design and developing a prototype we entered an intense testing phase that sought to determine how well our pump, vasculature, and blood models could mimic their human counterparts. The results of this testing phase are summarized below. They serve to demonstrate that our model is an effective, though imperfect, reproduction of the functional human circulatory system.

3.2 Pump Testing

3.2.1 Flow Rate Testing

When we set out to create our model of the cardiovascular system we needed to decide what aspects were most important in relation to traumatic hemorrhage. Most traumatic hemorrhage deaths are a result of hypovolemic shock when the patient loses blood very rapidly so the first parameter we decided to model was flow rate. We focused specifically on the flow rate through the arteries where someone experiencing traumatic hemorrhage would lose blood the fastest. We were able to successfully collect flow rate data to measure the accuracy of model compared to that of the human heart.

3.2.1.1 Flow Meter and Arduino Control

The options for a frugal flow rate meter that was compatible with our system design was very limited. We decided that the best option would be using an Arduino compatible flow meter. The flow rate data for our initial pump design was collected using a ½'' plastic NPS threaded Arduino liquid flow meter that was connected to an Arduino Uno board. The Arduino Uno board was coded to collect the flow in liters per minute every second. The flow meter was placed approximately 1' from the output of the pump to give adequate length of tubing to allow the turbulent flow to return to laminar flow for accurate readings.

3.2.1.2 Testing Results

After collecting and analyzing the data, we found that with our initial pump design was able to achieve a maximum flow rate of 10 liters per minute which is comparable to that of the heart which has a maximum flow rate of 8 liters per minute. Although this is not an exact match we decided that it was acceptable because we would rather over engineer than under engineer the parameters. Any hemorrhage solution that is created and tested on our system that can withstand the maximum flow rate of 10 liters per minute will also be able to handle the maximum flow rate of the human cardiovascular system which is 8 liters per minute. These values are referenced and compared in Tables 3.1 and 3.2 below.

Table 3.1 Flow Testing Results
Our model was characterized by a high degree of variability in flow rate even across large sample sizes. This is a testament to the imprecision of our handheld drill power supply.

Test	Avg. Flow Rate (lpm)	Standard Dev.	Sampling Points
1	4.92	0.89	37
2	6.25	1.47	57
3	4.20	0.65	92
4	5.23	1.15	47
5	6.78	1.24	58

Table 3.2 Comparison Between Flow Data and Heart Model
This table clearly demonstrates which properties of the human heart our model was able to replicate and which ones it was not. All of the values listed for the human heart come from literature, but are sometimes approximated from within a range of acceptable values.

	Our Model (N=5)	Human Heart (Lit)[35]	Valid Modeling?
Overall Avg. Flow Rate (lpm)	5.47	6.00	YES
Overall Standard Dev.	1.04	2.00	YES
Stroke Volume (ml)	23.00	80.00	NO
Average Pulse Rate (bpm)	240.00	80.00	NO
Maximum Pulse Rate (bpm)	440.00	240.00	NO

With the trigger control of the drill to control the pump, which will be discussed in depth in Chapter 5, we were able to achieve and average flow rate of 5.5±1.0 liters per minute, which is well within the range of the average flow rate of the heart which is 6.0±2.0 liters per minute. The flow data was averaged from 5 different runs, controlling for all possible variables with each run.

3.2.1.3 Improvements to Flow Data Collection

There were some difficulties in collecting flow data. The first noticeable difficulty was the frequency in which the flow rate was collected. Instead of reading the flow rate once every second, we wanted to collect the flow rate every quarter of a second or every half second. In traumatic hemorrhage, life or death can often be determined by a matter of seconds so our design

[35] "Normal Hemodynamic Parameters," LiDCO Group, 106. < http://www.lidco.com/clinical/hemodynamic.php >

team wanted data that was accurate to every quarter second. Unfortunately the flow meter and Arduino code was unable to read data at such a high rate. When the time interval in the Arduino code was changed to collect the flow rate at anything less than once per second, the flow meter and Arduino board was unable to collect any readings at all. This problem can be attributed to the low processing power of the Arduino board which is made to control simple electronic circuits. In order to collect flow data at an increased rate a higher processing control board must be used along with a more accurate meter. Unfortunately using more powerful equipment also comes with a higher cost which we may or may not be able to accomplish.

3.2.2 Pressure Testing

The second parameter that we decided to control for and model our system to match that of the heart was the pressure of the system. The cardiovascular system of the human body has a blood pressure built up from the pumping of the heart, the elasticity and the resistance of the blood vessels. The average systolic blood pressure of the human body is 120 mmHg or 2.32 psi, while the upper range of blood pressure is around 200 mmHg or around 4 psi. However we were not able to collect accurate data or replicate these parameters within our system for a various number of reasons.

3.2.2.1 Improvements to Pressure Sensor and Pressure Sensor Hub

Since we were using the Arduino system to control the flow meter, with future plans of using the Arduino system to control the whole pump, it made sense to also use a pressure sensor that was controlled by the Arduino system. The pressure sensors available were also very limited and incorporating them into a fluid system proved to be very difficult. The pressure sensor we decided upon has very short electrode legs meaning that the breadboard that it was connected to needed to be very close to the fluid system. This posed a great risk in shorting out the entire electrical system if any fluid came in contact with live electronics. With those constraints in mind the part of the sensor that came into contact with any of the fluid needed to be waterproof. Using a PVC and plastic connector we incorporated the pressure sensor into the pressure sensor hub using epoxy. Unfortunately because the surface that the sensor was epoxied to was not flat it prevented accurate readings from the sensor. The fluid in the system was able to push underneath the sensor and prevent the sensor from accurately reading the correct pressure on the tubing walls.

To improve this pressure system we would need a new pressure sensor hub in which the sensor is able to lay on a flat surface to more accurately measure the pressure that the flowing water is putting on the walls of the tubing. Once we have a flat surface to anchor the pressure sensor to, then the fluid will no longer be able to seep under the sensor and the sensor will be able to obtain accurate readings. To improve the sensor itself, we would need to extend the length of the electrodes to remove the possibility of the rest of the electronics being exposed to the fluid. This can be completed in various ways, but the most immediate and frugal way would be to solder electrode extenders to the pressure sensor in order to connect them better.

3.2.3 Conclusions

We were able to accurately and definitively confirm that our initial pump design was able to model the flow rate of the blood through the arteries. This parameter was the one we really focused on and is the most important aspect to consider when analyzing the traumatic hemorrhage. We were unfortunately unable to obtain accurate pressure data, but with time we could both improve the frequency of data collection for the flow rate as well as obtain accurate pressure data.

3.3 Vasculature Testing

3.3.1 Material Property Testing

In order to validate our choice of vasculature material, we performed mechanical properties tests on our chosen material, silicone, and another high-quality candidate material, PVC vinyl. The test we selected was a test of tensile strength. This kind of testing can, if completed, yield accurate values for a material's Young's Modulus which is a measure of that material's elastic properties.

3.3.1.1 Tensile Strength Testing Protocol

The tensile strength testing we conducted using the Biomomentum Mach-1 Mechanical Testing System. We took six samples for silicone tubing and six samples for vinyl tubing. Three samples of each material was 2" x 0.25", and three samples were 2" x 0.125". The Mach-1 was set up and calibrated for tensile testing using a 10 g calibration mass. After calibration, the samples were loaded one at a time into the system by using the clamps to secure the silicone into the base and mobile arm of the system. The system was manually moved up until the sample had no slack. The program was set up for linear tensile testing, which measures the force needed to strain the material a certain distance. The stretch distance was set to 250 mm at 10 mm/sec. The machinery we used to perform tensile tests is diagramed in Figure 3.1 in sub-section 3.3.1.2.

3.3.1.2 Silicone Tubing Testing

Ideally, the 250mm distance would have been long enough to strain the silicone until the fracture point in order to see the complete plastic and elastic regions to calculate ultimate tensile strength, toughness, and resilience. However, the Mach-1 did not allow for enough strain to break any of the samples, so our samples only experienced elastic deformation during the test. While not complete, this data is still significant and allows us to calculate a Young's Modulus for the material and compare it to that of human arteries and other silicone tubing found in literature.

Figure 3.1 Tensile Testing Machinery
The image on the left shows the Mach-1 Mechanical Testing System by Biomomementum that was used for all our tensile testing. The image on the right shows a sample of polyurethane foam being stretched during the tensile testing process.

The Young's modulus we calculated was significantly higher than that found in other studies of similar tubing, which could be attributed to different thickness of the silicone sample used. Silicone is significantly more elastic and stronger than human arteries, but it still provides a viable elastic model for testing traumatic bleeding. Testing results are summarized in Figure 3.2.

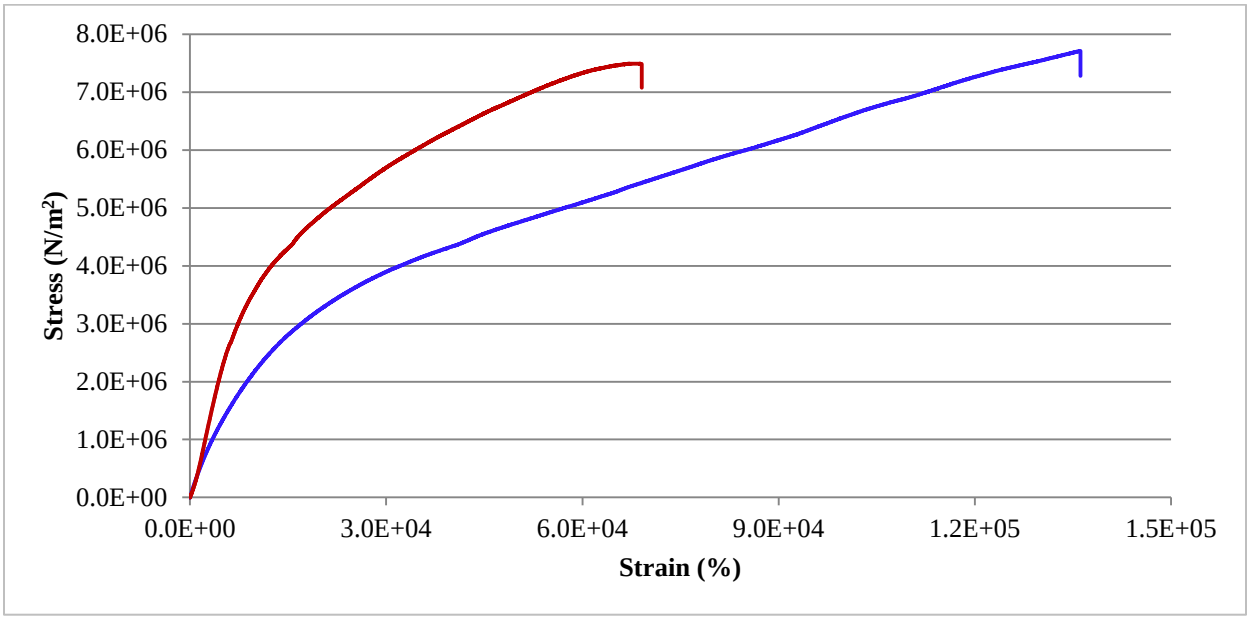


Figure 3.2 Stress vs. Strain Curves for Silicone and Vinyl
These curves represent the data collected during our tensile testing of samples of silicone tubing (curve shown in blue) and PVC vinyl tubing (curve shown in red). The slope of the curves during their most linear stretches can be roughly interpreted as equivalent to the respective Young's Moduli of the material (though slope values are off). What is important to note is how much more elastic silicone appears to be then vinyl, and how much further it can be stretched under the same force as a result.

3.3.1.3 Vinyl Tubing Testing

The vinyl tubing was found to exhibit a much higher young's modulus, but also a much lower yield strength. Vinyl experienced some plastic deformation during our tests while silicone tubing did not reach the plastic region at all. The larger elastic region of silicone works better than vinyl in two ways. One, it provides a better match for the arteries that we are trying to mimic, and two it allows for greater rebound in the peristaltic pump. The pump quickly and continuously compresses the tubing to push flow through the system, so if the material is not elastic enough, it will not be able to restore its original shape and not be able to push as great a volume through our system as quickly.

3.3.2 Fluid Dynamics Testing

Care was taken when designing the branched vasculature model to ensure that the fluid dynamics within the system would be as accurate to reality in the body as possible. Ordinary blood flow in the body varies consistently between laminar and turbulent in nature, which means that blood is constantly alternating, depending on its location, between a uniform and parabolic flow pattern and a somewhat randomized, less efficient flow profile.[36] It would have been very difficult to predict and recreate the range of ordinary fluid profiles that will be achieved throughout the circulatory system.

It was possible, though, to virtually determine what fluid profiles would likely be achieved in our model under some assumed flow conditions. From the expected properties of our blood substitute and the expected flow rates of our cardiovascular model we were able to determine projected Reynold's numbers for the flow in each level of our system. These numbers are summarized in Table 3.3 immediately after this paragraph. In order to calculate these numbers we assumed even division of flow at each vessel bifurcation. We also ignored the short distance between vessel bifurcation and the diameter drop that was necessary as a result of coupling and part limitations. These assumptions are reasonable – without them, any fluid analysis would be extremely difficult.

Table 3.3 Reynold's Numbers for Flow in Various Parts of Vasculature Model
Notice that SI units are used to simplify the calculations and thus that the first 0.01270 m pipe is the main 0.5" diameter outlet pipe of the pump, the second is one of the 0.5" diameter pipes after the first split, the 0.009525 m pipe is one of the 0.375" diameter pipes and so on. Note also the exceedingly high Reynolds Numbers (Re).

Diameter (m)	Flow Rate (m^3ps)	Velocity (m/s)	Viscosity (m^2/s)	Density (kg/m^3)	Re
0.01270	9.167E-05	28.49	0.0044	1060	87000
0.01270	4.583E-05	14.24	0.0044	1060	44000
0.009525	2.292E-05	12.66	0.0044	1060	29000
0.006350	1.146E-05	14.24	0.0044	1060	22000
0.003175	5.729E-06	28.49	0.0044	1060	22000

Reynold's number is very useful for characterizing flow through pipes in fluid dynamics. Typically a Reynold's number under 2000 indicates laminar (parallel and unidirectional) flow while a Reynold's number higher than this indicates turbulent flow. The Reynold's numbers for

[36] Ku, David N., "Blood Flow in Arteries," *Annual Review of Fluid Mechanics*, 29, 1997. 399-434.

our device are incredibly high, indicating with near certainty that our fluid would experience turbulent flow throughout the model. This is likely because our device does not simulate smaller blood vessels and also fails to branch flow enough to simulate the circulatory system. Exponential branching in the body is characterized by a higher order exponentiation than our model.

3.3.3 Conclusions

Our vasculature model could be improved upon by more accurately replicating the arteries, but what we currently have is complete and accurate enough to allow new hemorrhage solutions to be tested upon it.

3.4 Blood Mimicking Fluid Testing

3.4.1 Newtonian Viscosity Testing

3.4.1.1 Initial Viscosity Testing

Our initial blood mimicking fluid recipe, as discussed earlier, called for a mixture of water (which served as the liquid base), glycerol (which served as a viscous thickening agent), and xanthan gum (which served as a shear-thinning agent). This recipe came from the scientific community and was determined during our literature search, but we needed to perform testing to determine the relative amounts of these agents required to make our blood substitute.

Based on literature suggestions and recipes we determined that the best method by which to characterize the substance ratios would be to use percentage by weight descriptors. We also determined from literature that an ideal xanthan gum concentration would be around 0.015 wt%. In order to determine the relative concentrations of glycerol and water needed we performed a series of dynamic viscosity tests in which we varied the glycerol concentration from 0 – 20 wt%.

We performed our tests using a U-tube Ostwald Viscometer. This form of viscosity testing measures the time it takes a fluid of unknown viscosity to drain through a carefully calibrated section of glass pipe between two marks. Three separate tests were performed for each fluid recipe yielding three separate times. These times were then averaged and translated into viscosity measurements using the pre-supplied constant associated with the viscometer we used. The resulting value yields an accurate approximation for the dynamic viscosity of the recipe in question. These results are summarized in Table 3.2 below, and images of our team performing the testing can be found in Figure 3.4 shortly after.

Table 3.4 Dynamic Viscosity Testing of Initial Blood Mimicking Fluid Recipes
Note that the average time value is calculated for N = 3. The Tube number indicates the size of the viscometer and is a dimensionless label. The K value is the viscosity calculation constant associated with a certain Ostwald Viscometer U-tube. The final viscosity value of 191.80 is clearly an outlier – no recipe would include or call for a glycerol concentration that high.

Sample	Glycerol (wt%)	Avg Time (s)	Tube #	K (mm^2/s^2)	Dynamic Viscosity (cP)
1	0.00	88.94	150	0.035	3.17
2	3.75	93.95	150	0.035	3.33
3	7.50	106.93	150	0.035	3.82
4	15.00	126.52	150	0.035	4.61
5	30.00	191.47	150	0.035	7.24
6	60.00	659.68	300	0.250	191.80

These values, with the exception of those calculated during test 6 (omitted as being beyond the linear or usable range of values for out testing) were then plotted in graphical form and linearly regressed to determine their relationship. The resulting graph is shown in Figure 3.3 below.

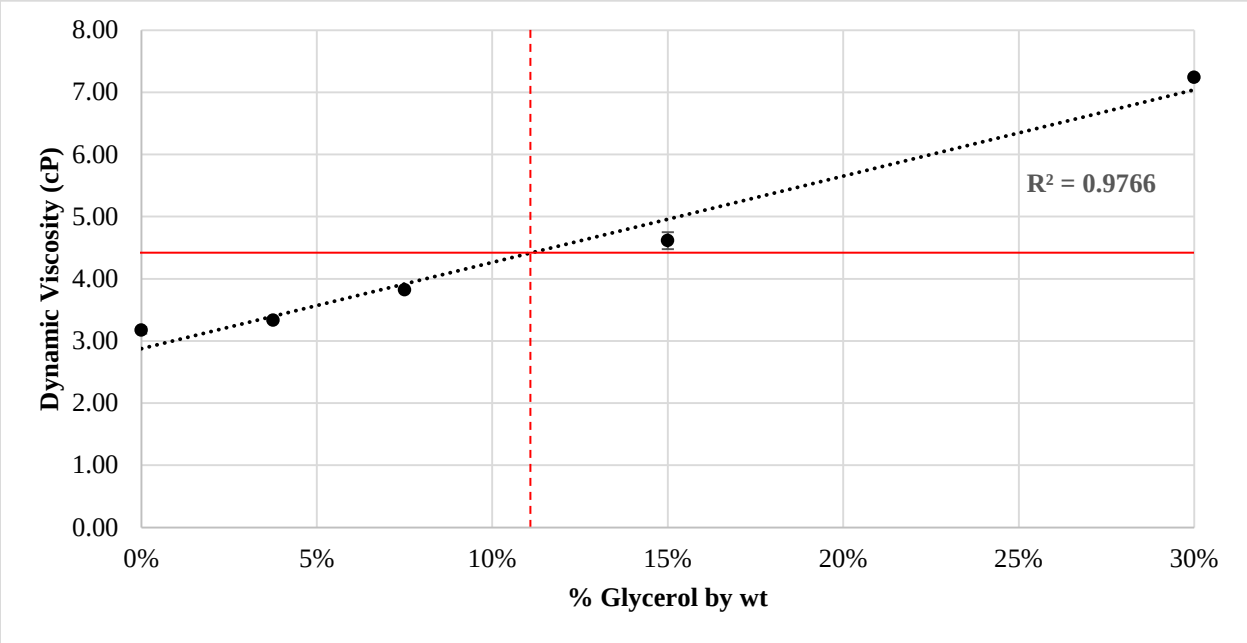
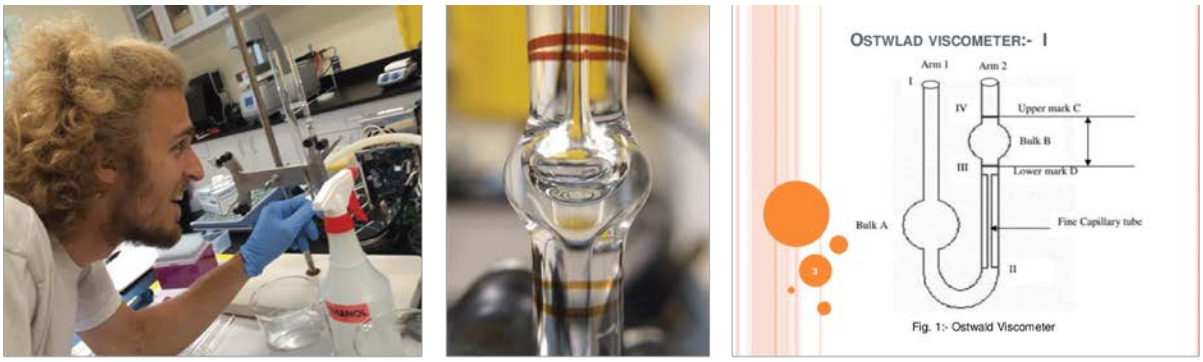


Figure 3.3 Dynamic Viscosity Testing and Linear Relationship of Initial Blood Mimicking Fluid Recipes
The figure shows the relationship between dynamic viscosity and glycerol concentration in our blood mimicking fluid. The horizontal red line indicates the desired viscosity of 4.4 cP. The vertical red dashed line indicates the expected concentration of glycerol required to yield this viscosity. Note that this concentration of Glycerol appears to be right around 11.5 wt% although any concentration between 10 and 15 wt% would fall within the normal variance range for blood viscosity.



(A) (B) (C)

Figure 3.4 Viscosity Testing Images
(A) Nic in the labs performing a viscosity test (B) A close up of the timed drain segment of our Ostwald U-Tube Viscometer. The fluid is raised to a level above the top line and then allowed to fall naturally. As it passes the first line the timer is started; the timer is not stopped until the fluid level drops below the lower line (C) A more explanatory diagram of an Ostwald Viscometer. Image Courtesy of Tenison Basumatary, *Gauhati University*, 2015.

3.4.1.2 pH Adjusted Recipe Viscosity Testing

As discussed in the previous chapter, we decided to attempt to replicate physiological pH with our final blood substitute recipe. To this end we switched our primary liquid base media from water to a phosphate buffer with pH near 7.4. In order to validate this new recipe, we generated a matrix of new recipes (based on different glycerol and xanthan gum ratios) and performed more viscosity tests on them.

The matrix of recipes we created varied the xanthan gum concentration from 0.010-0.020 wt% and the glycerol concentration from 5-20 wt%. Because our previous recipes indicated an ideal glycerol concentration of 12.5 wt% and an ideal xanthan gum concentration of 0.015 wt% we thought these ranges would capture the ideal blood substitute recipe even with the new phosphate buffer base. We did not anticipate that the phosphate buffer would significantly affect the relationship between glycerol and xanthan percentage by weights and dynamic viscosity.

We were unfortunately wrong. The pH replicating phosphate buffer had a dramatic effect on the viscosity measurements taken. As the data in Table 3.5 below indicates, no recipes we tested reached the 4.4cP viscosity benchmark we had set in spite of our attempts to bracket previous successful recipes. We have not, as yet, had enough time to pursue another batch test with higher concentrations – until we perform such a test we cannot validate our pH replicating blood mimicking fluid model. We can, however, extrapolate from trendiness produced from our data. These trend lines and a graphical representation of our data, can be found below in Figure 3.5.

It appears from this graph that the trend line corresponding to the xanthan gum concentration of 0.020 wt% would be the first to intersect with the target viscosity. This would occur at a glycerol concentration of somewhere around 24.00 wt%. Obviously, it is improper to draw conclusions from improperly (i.e. extrapolated) results, and so we have chosen to refrain from drawing conclusions relative to our pH substitute at this juncture. We choose instead to supply what we feel would likely be the ideal recipe (shown in Table 3.6) along with the supposition that it would probably work well. Ultimately, we need to perform more testing to confirm this recipe.

Table 3.5 Dynamic Viscosity Testing of Phosphate Buffer Blood Mimicking Fluid Recipes
Each colored division indicates a different constant percentage of xanthan gum. The darkest gray (at the top of the table) represents a constant xanthan concentration of 0.020 wt%. The lighter gray indicates a concentration of 0.015 wt%. The white boxes indicate a constant xanthan concentration of 0.010 wt%. Sample 12 could not be tested. The tube column has been omitted from this table as all tests were run in a #150 Ostwald Viscometer with the indicated K value.

Sample	Glycerol (wt%)	Xanthan (wt%)	Avg Time (s)	K (mm^2/s^2)	Dynamic Viscosity (cP)
1	4.97%	0.0210%	67.84	0.035	2.39
2	10.72%	0.0213%	81.76	0.035	2.94
3	15.30%	0.0210%	90.40	0.035	3.29
4	20.32%	0.0211%	112.15	0.035	4.12
5	5.24%	0.0158%	62.32	0.035	2.21
6	10.15%	0.0159%	63.80	0.035	2.29
7	15.12%	0.0160%	80.27	0.035	2.93
8	20.00%	0.0159%	90.40	0.035	3.36
9	5.12%	0.0106%	54.57	0.035	1.94
10	10.07%	0.0106%	66.39	0.035	2.39
11	15.13%	0.0107%	74.50	0.035	2.71
12	DNF	DNF	DNF	DNF	DNF

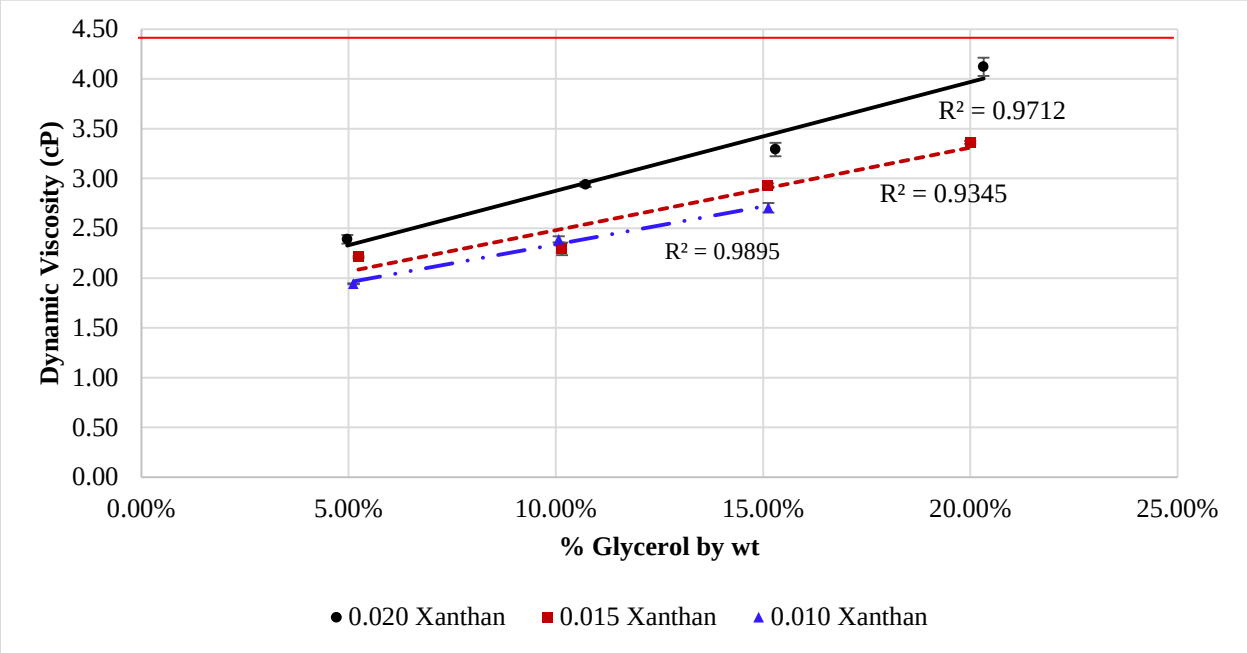


Figure 3.5 Dynamic Viscosity Testing and Linear Relationship of pH Replicating Fluid Recipes
The figure shows the relationship between dynamic viscosity and glycerol concentration in our phosphate based blood mimicking fluid. All three standardized concentrations of xanthan gum have their own data series and trend line so that the relative effects of varying both concentrations can be observed. The red horizontal line still indicates the target viscosity of 4.4 cP – a goal woefully out of reach here. Note the missing data point within the xanthan = 0.010 wt% series.

Table 3.6 Recommended Final Blood Mimicking Fluid Recipe

Material	Wt%	Purpose
Phosphate Buffer	75.980	pH 7.4 Buffer and Liquid Base
Glycerol	24.000	Viscous Thickening Agent
Xanthan Gum	0.020	Shear-Thinning Viscosity Modifier

3.4.2 pH Testing

Before we conducted the viscosity experiment, we measured the pH of the 11 samples we mixed. We first calibrated the pH probe using a basic, neutral, and acidic calibrating buffers to ensure accurate readings. We then submerged the pH probe into 5 ml of the solution in a 10 ml centrifuge tube, waited for the read-out to stabilize, and recorded the value. Between each sample the probe was rinsed with DI water. The pH for each sample is recorded here in Table 3.7. Summary statistics for these data are presented in Table 3.8 immediately after.

Table 3.7 pH Values for Previously Created Blood Mimicking Fluid Recipes
Note that, as before mentioned, sample #12 was not testable.

Sample	Glycerol (wt%)	Xanthan (wt%)	Dynamic Viscosity (cP)	pH
1	4.97%	0.0210%	2.39	7.266
2	10.72%	0.0213%	2.94	7.257
3	15.30%	0.0210%	3.29	7.270
4	20.32%	0.0211%	4.12	7.243
5	5.24%	0.0158%	2.21	7.241
6	10.15%	0.0159%	2.29	7.253
7	15.12%	0.0160%	2.93	7.272
8	20.00%	0.0159%	3.36	7.256
9	5.12%	0.0106%	1.94	7.254
10	10.07%	0.0106%	2.39	7.257
11	15.13%	0.0107%	2.71	7.241
12	DNF	DNF	DNF	DNF

Table 3.8 Summary Statistics for Table 3.5
Note that our samples have a significantly different average pH than the human population average down to the Alpha = 0.0001 level and below. This indicates an extremely significant difference that is unlikely to have arisen by pure chance.

Target pH[37]	7.390
pH Sample Mean (M)	7.255
Sample Standard Deviation (s)	0.011
Sample Number (N)	11
Significance	Alpha = 0.0001

[37] "Table 1. Summary Statistics for the Distribution of pH and (H+) Values in Blood from Humans and Lobster (*homaris Vulgaris*)", in *Regulation of Tissue pH in Plants and Animals: A Reappraisal of Current Techniques*, ed. S. Egginton, Edwin W. Taylor, J. A. Raven, New York: Cambridge University Press, 1999. 356.

3.4.3 Conclusions

While we have not developed an ideal blood mimicking fluid at the time of this writing, we have the data and trends to suggest an ideal recipe for the glycerol, xanthan gum, and phosphate buffer needed for our fluid. With slight modifications to the pH of our phosphate buffer we can achieve the proper pH and with a slight change in the concentration of glycerol and xanthan gum we can achieve the proper viscosity.

3.5 Solutions Testing

3.5.1 Solution Candidate Selection

Like the process of searching for a solution for any type of problem, we had to start with a broad scope and slowly narrow down the possible solutions. We started off our solution candidate search by selecting a wide variety of sealants and foams that we believed had the physical properties to stop traumatic hemorrhage, or in the case of our model, seal or plug an “injured” part of our tubing system. We decided to test a variety of different material solutions including, silicone sealant, insulation foam, epoxy, J.B Weld brand epoxy, and polyurethane foam.

Although none of these solutions can actually be used in the human body because they would be toxic, we wanted to focus on the physical characteristics of the possible material solutions. If the material were able to pass all of our screening tests as well as seal or plug an “injured” section of our tubing system, then we would be able to note the physical characteristics of the solution and use them as constraints in developing a biocompatible material.

3.5.2 Preliminary Solutions Testing

The preliminary testing that we ran our material solutions through were completed for a number of different reasons. The first and obvious reason was to find the physical characteristics which helps solve traumatic hemorrhage. The second reason was to confirm that we have a viable platform that is injurable and replicates some of the important parameters of the heart and human cardiovascular system.

3.5.2.1 Waterproof Testing

The first test that we put the materials through was waterproof testing. The purpose of this test was to see if the materials chosen would be able to plug the bottom of a section of tubing and hold 2 ml of water without letting any water leak through the bottom. The tests were run at 3 different time intervals (5 minutes, 2 minutes, and 30 seconds) to allow a short amount of time for the material to set. The first test was run at 5 minutes. This means the material was placed

and plugged the bottom of the tubing, then allowed to set for 5 minutes before 2 ml of water was added.

For both the 5 minutes and 2 minutes test, all of the material solutions were able to hold 2 ml of water without letting any of it leak through the bottom of the tubing. However, for the 30 second time interval test only the silicone sealant as a well as the insulation foam were able to hold 2 ml of water without letting any leak through the bottom of the tube. Both types of epoxy as well as the polyurethane foam were not able to set and harden quickly enough to hold 2 ml of water in a small section of tubing. Figure 3.6 shows the results of this waterproof test.

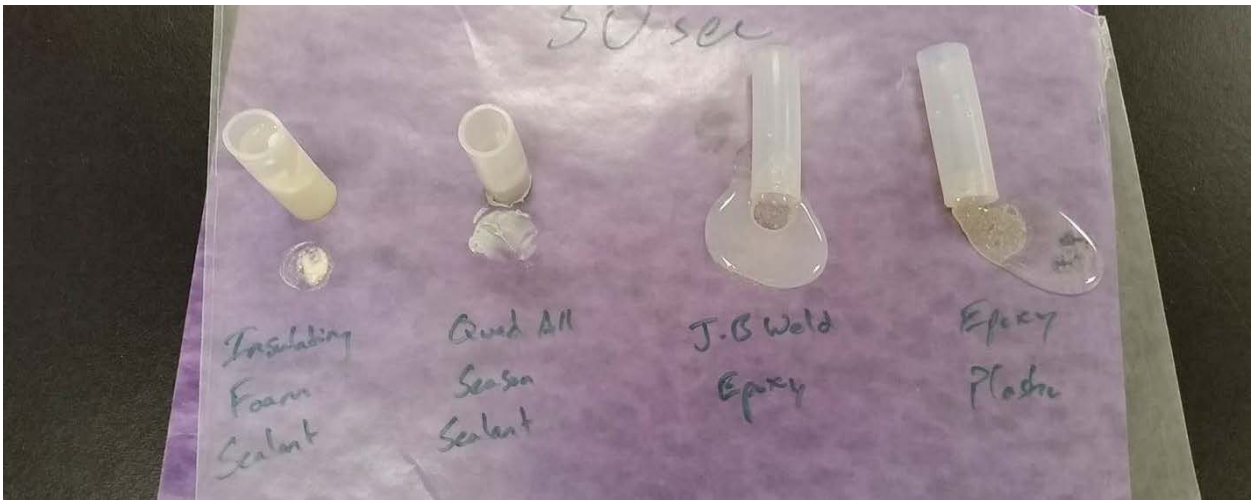


Figure 3.6 Waterproof Solution Testing
This image shows the solutions after they were allowed to set in the bottom of a piece of vertically positioned silicone tubing. After 30 seconds of set time, 2ml of water was added to each of the tubes and this tube lifted from the wax paper (hence the distinctive rings of material left behind be the leftmost two tests).

3.5.2.2 Live Flow Testing

Once we narrowed down the possible solutions to the silicone sealant and the insulation foam, we decided that the next screen for testing our possible material solutions was a live flow test. During this test we plugged one of the two 0.5” tubing sections coming off of the Y-splitter with our possible material solutions. We then ran our pump to see if the possible material solution would hold up against the pressure that the pump provided. In our test we left one of the two 0.5” tubing sections open in order to replicate the conditions of arterial bleeding in the extremities. When one blood vessel in the extremity is blocked, that extremity is still able to achieve circulation because there are multiple blood vessels leading to the same area.

The silicone sealant was viscous and strong enough to hold up against the pressure that the pump provided and all of the fluid flowed through the open 0.5” tubing and into our output reservoir. The insulation foam’s physical properties did not allow it to hold up against the pressure that the pump provided and the foam was pushed out of the tube with the fluid.

3.5.2.3 Injected Material Testing

We narrowed down our possible material solutions only the silicone sealant. Our final test was an injected material test. This test consisted of running the pump and as fluid was flowing through the system. We cut a small section of easily replaceable tubing to replicate an arterial bleed and then took the silicone sealant and injected it into the tubing at the site of injury. The silicone sealant stopped the flow of liquid through the tubing for a few seconds as well as sealing off the injury so that no additional fluid was able to leak out of the tubing. The liquid then pushed through the silicone sealant so that flow was able to continue through the injured section while the injury to the tubing was sealed by the silicone.

3.5.3 Conclusions

The qualitative data that we were able to collect from the solutions testing is the beginning of the material solution research. The physical properties of the silicone sealant make it an excellent candidate to model future biocompatible solutions with. However, the physical properties of the sealant are specific to only one type of arterial bleeding, where we are assuming that the artery that is damaged is still partially intact and not completely severed. If the artery were completely severed, a material with different physical properties would be needed. The sealant is malleable enough to seal off the tubing but allows the pressure of the fluid to push the sealant to the walls of the tubing and allow the fluid back through. For an injury where the artery is actually completely severed, a biocompatible material that is injectable and is able to occlude the blood vessel while withstanding the pressure of the fluid is needed.

Future Considerations

4.1 Pump, Drive, and Pump Housing

4.1.1 Pump Design

Our final pump design is currently nonfunctional for a variety of reasons. A lot of these have to do with failures in the electrical control system, but some of them have to do with the pump itself. We have planned a number of potential improvements to our pump concept that seek to mitigate these problems and improve functionality.

4.1.1.1 Improvements to the Pump Size

When we set out to design a pump to replicate the heart in the human cardiovascular system, we had initially hoped to replicate more of the heart's parameters than we were ultimately able to with the final pump size we selected. We even purchased some materials to this end. The problem with the current pump's size is that it does not allow us to replicate the stroke volume of the heart which is around 4.27 in^3 in most adults.

The stroke volume provided by our pump, as discussed earlier, is the interior volume of the length of tubing stretched between two of the pumps rollers. This is mathematically a function of the length and diameter of the tubing as well as the diameter of the cake pan housing, the number of rollers used, and the degree angle between the rollers themselves. Ultimately, by choosing a 4 roller design that operated within a 9" cake pan housing lined with 0.5" silicone tubing, we limited our maximum stroke volume to around 1.37 in^3.

We have the materials required to increase this number. If we wanted to over engineer, using a 3 roller design, a 12" diameter pan, and 0.75" diameter silicone tubing we could achieve a maximum stroke volume of around 5.55 in^3 - near enough to the target human value to adjust downwards. This would require relatively few design modifications to the pump itself other than a redesign of the central manifold to accommodate 3 evenly spaced rollers. In the unlikely event that we chose to under engineer our pump, we could strike even closer to the 4.27 in^3 target. Keeping with our four roller design, but switching to the larger 4" diameter pan and 0.75" diameter tubing, we could achieve a maximum stroke volume of 4.16 in^3.

4.1.1.2 Improvements to the Central Manifold

The appearance of the pump's central manifold was improved significantly after our decision to precision mill the central hub out of HDPE plastic, but some more functional and less cosmetic

improvements are still required. It would be a long term goal of ours to mount nicer rubber coated rollers onto spring loaded mounts so that they would maintain constant force on the tubing as they passed around it. This would also reduce the amount of vibration passed into the pump and pump housing as a result of the imperfect rotation of the manifold (like a suspension system).

4.1.1.3 Improvements to the Primary Housing

Currently our pump is constructed within a spring-form cake pan. This is hardly the most aesthetically pleasing or professional method of construction. Future improvements would include precision milling a housing ring, bottom, and top out of reinforced transparent acrylic plastics. This would improve the sturdiness and durability of the pump, reduce the amount of dust and contaminants that enter the machinery, and still allow visualization of the function and design. It would also allow for material failures and breaks to be easily identified and allow easy repairs to be made.

4.1.2 Drive System

Our current pump is supposed to be operated by a remotely controlled electric motor positioned directly underneath the pump in a different housing. This electric motor was removed from a battery operated drill. Unfortunately, because of its origin, we have not been able to effectively power the motor chosen thus far. Drill motors are carefully design to operate based on a specific, and often proprietary, power supply. In our case, the electric motor we used was meant to work with a 12 V rechargeable battery on a DC circuit. We were trying to power this motor with 120 V AC power provided by a standard receptacle. This necessitated the use of a power supply/switch to function as an intermediary between the device and the wall outlet. We attempted to borrow one from another Bioengineering project, but have not been able to calibrate it correctly thus far.

Going forward, we see two options for improvement in the drive system itself. The first involves selecting a different motor: perhaps one with a stepper function like that of a drill motor but that would run on 120 V AC power or that would be more easily powered from a standard power supply/switch. Certainly we would look for motors that do not have proprietary electronics associated with them. The other option would be to purchase a better power supply that fits the parameters of the motor we are currently working with. It is likely that the best solution would be to buy a new combination of both - a motor and power supply designed to function as well together as the original drill motor and drill battery were.

4.1.3 Housing Design

5.1.3.1 Housing Design

Currently our pump support housing is constructed from 6 panels of 0.25” material joined together loosely with box joints into a cube shape. The pump is nested into a series of large cuts in the top of the box. There are a number of improvements we would like to make to this design. We would like to prevent the pump from overhanging any part of its housing to improve the

stability of the final design. Whether this should be accomplished by increasing the box footprint or manufacturing a round housing is not currently clear. The box/housing should also be made heavier towards the bottom to increase its center of gravity and resistance to motion upon vibration. More space could also be allowed in the bottom of the device for control electronics.

4.1.3.2 Housing Material

The box is currently constructed out of a softwood plank 0.25” thick that is easily cut with a laser cutter. This material is relatively flimsy and is unlikely to hold up under the long term stress it would undoubtedly incur in a testing setting. We would like to brainstorm new materials from which to construct this box and housing. It seems unlikely that metal would be an efficient choice with the possible exception of something strong and lightweight like aluminum or titanium (but the expense alone would render these unusable). Strong plastics seem ideal, particularly strong plastics that are easy to work with, drill, and machine. HDPE or acrylic could be excellent choices for future models and modifications. These could be joined with screws or bolts and vibration resistant putty.

4.2 Vasculature Model and Injuries

4.2.1 Vasculature Layout

We are currently very satisfied with our vasculature model layout, but there is always room for improvement. Currently, our model consists of an exponentially branching network of tubing that decreases in diameter linearly at every branch. The human vasculature network does not branch in nearly as neat a fashion, and so one improvement to our model could be to develop multiple vasculatures each with a different branching pattern. Additionally, we could alter the criteria for stepping down the tubing diameter so that decreases in diameter happen less regularly.

4.2.2 Tubing Sizes

There are two main improvements and a few smaller ones that we would like to make to our vasculature model regarding tubing size. These improvements affect the beginning and end tubing sequences of the arterial model. The first level of tubing our model involves is 0.5” in diameter, as mentioned earlier. This eventually branches 4 times and drops in diameter 3 times before the final pipes, each 0.125” in diameter, are reached. The largest blood vessel in the body, the Ascending Aorta, is nearly 1” in diameter. This is dramatically larger than the largest blood vessel our model simulates. Additionally, human arteries and arterioles drop to diameters much smaller than 0.125”.

We would like to improve our model to reflect these key facts going forward. Our future vasculature design would begin with 1" diameter silicone tubing and branch to 0.0625" diameter tubing if such tubing is available. If not, the smaller vessels are not as concerning or important to the model as the larger ones. It would also be our goal to introduce more diameter reduction steps in the series to lengthen our overall model perhaps stepdown options of 0.0625" are available. It would also be nice to purchase tubing for our model that more accurately reflected human artery wall thickness. Though the silicone tubing comes close, it is still too thick to be an accurate substitute for the body's vessels.

4.2.3 Capillary Modeling

Our model currently does not feature any arteriole or capillary modeling component. One of the benefits of the these smaller vessel networks in the body is that they produce the vascular resistance that allows blood pressure to build to the levels required to move blood throughout the body and against gravity. Our model currently offers very little vascular resistance – fluid is never forced into such small tubes that it backs up against the flow. Adding such a component to the system would improve our ability to model blood pressures more accurately.

It is clearly beyond the scope of our project to model the smaller vasculature with actual tubing, but it would not be impossible to create vascular resistance in other ways. We have envisioned an adjustable box chamber that contains swappable screens or porous filters as a method of creating some flow resistance. This box chamber would fill with fluid and then begin to impede flow much in the same way that silica gel impedes flow in gel column chromatography.

4.2.4 Venous Modeling

Our model is also noticeably lacking in proper venous system simulation. Currently the venous system is modeled by a 5 gallon bucket and a large diameter silicone tube. It is adequate, but hardly aesthetic, accurate, or professional. At the very least our design requires some refinement to more adequately replicate the human Vena Cava and blood storage system. A transparent bucket with a lid would be a good start, but ultimately a more professionally designed reservoir system would improve our design.

4.2.5 Injury Simulation

The current method of injury simulation we are using involves inflicting trauma directly onto fully exposed silicone tubing. This, of course, is not how injuries occur in animals or humans in reality. The arteries and veins of the body are never fully exposed to the outside and injuries are rarely as easy to identify as we have made them seem to be. Severe bleeding does not always occur at the actual site of trauma to the blood vessel in question. Often the blood vessel will retract deep into the soft tissue – a prime example of this can be seen in the movie *Black Hawk Down*.

Our model could more accurately simulate bleeding injuries if the injurable sections were surrounded by a material that simulated the viscera and soft tissue. In our research one of the best ways to simulate these tissues is using raw meat samples or animal carcasses. Our labs at Santa

Clara are not well-equipped to handle these kinds of biological samples, but it is possible that non-biological substitutes from rubber to hydrogels could function in this capacity. It is worth exploration and further study.

4.3 Blood Mimicking Fluid

4.3.1 Shear Thinning Rheometry

We were unable to analyze the shear-thinning properties of our viscous blood substitute before our time with this project expired, but we have two future methods of testing this property to recommend to future researchers.

4.3.1.1 Rotational Rheometry

The traditional laboratory method of testing a substances shear properties involves rotational rheometry. This process uses a pair of rotating discs separated only by a thin layer of the substance to be evaluated to determine its viscosity under different shear forces. If the viscosity changes as the force exerted on the substance changes, the substance demonstrates non-Newtonian shear properties. The machines required to test these properties are very expensive and complicated to operate. Despite numerous attempts to secure one for our research, we were unable to test on one of these complicated rheometers. Future testing on this machine is required before our blood mimicking fluid can be adequately validated.

4.3.1.2 Poiseuille's Law Rheometry

There is a cheaper alternative to rotational rheometry that we also considered. Poiseuille's Law is a well-known law of fluid dynamics that relates viscosity to a number of other fluid flow properties including vessel radius, flow rate, and fluid pressure. The equation is reproduced here:

$$Q = \frac{\pi P r^4}{8\eta l} \tag{4.1}$$

This equation can be used to determine the viscosity change in a fluid that is subjected to different forces – essentially it can be shoehorned into becoming a good test of shear properties. If fluid is forced through a narrowing pipe at a constant flow rate, the change in pressure between a point prior to the narrowed section and a point after the narrowed section is proportional to the change in the fluid's viscosity between these two points. A diagram of this test is shown here in Figure 4.1:

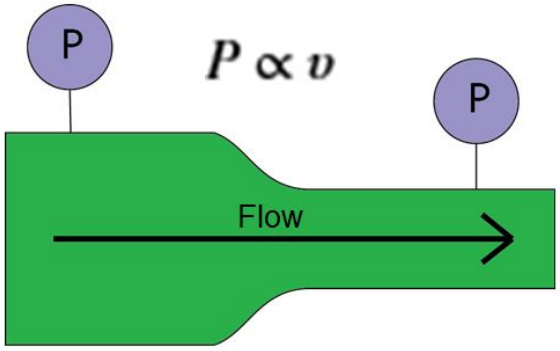


Figure 4.1 A Diagram of Poiseuille's Fluid Test
The green shape represents a pipe with a narrowing diameter. The two blue circles represent pressure gauges at two points on the pipe.

We were unable to perform this test because the pressure monitoring equipment was expensive and did not integrate easily into our design. Even the primary pressure we purchased for our device experienced numerous technical difficulties throughout our project. In the future we would like to have this test performed to corroborate or substitute for the more formal rotational rheology process as we attempt to validate our blood mimicking fluid.

4.3.2 pH Refinement and Physical Property Simulation

In order to refine the pH, it is simply a matter of modifying the choice of phosphate buffer from a 7.2 pH buffer used in our experiment to a 7.4 pH solution. However, as we found in our experimental data, changing the base from water to phosphate buffer lowered the viscosity at the same levels of glycerol and xanthan gum considerably. This decrease in viscosity must be accounted for in developing the experimental procedure.

Another direction to take would be exchanging the phosphate buffer, consisting of NaOH and KH_2PO_4, to phosphate buffered saline, consisting of KH_2PO_4, Na_2HPO_4, NaCl, and KCl. Phosphate buffered saline, or PBS, is a common solution used in cell culture and other biological experiments as a solution that mimics both the physiological pH as well as the physiological concentrations of salt, primarily sodium and potassium. The addition of the physiological salts would add an additional property of blood to provide an even more ideal blood mimicking fluid. Again the effect of the added saline might chemically interact with the other components of the colloidal mixture, altering the viscosity.

4.3.3 Temperature Considerations

We have assumed for our current blood mimicking fluid that experiments will all be taking place at 20 C, room temperature, in order to simplify testing and increase throughput. Our blood mimicking fluid at room temperature currently matches the properties of blood at 37 C. However, if a researcher using our cardiovascular platform would want to use our model to test a temperature sensitive polymer or test the effects of temperature on their material, it would not be possible with the current blood mimicking fluid. With a few simple modifications to the blood

mimicking fluid, it could most likely be used in a temperature-controlled cardiovascular model. The main concern would be viscosity, given that viscosity is temperature dependent. For example, the work of Nian-Sheng Cheng suggests a strong mathematical relationship between the temperature and viscosity of glycerol given the following equation:[38]

$$\mu_g = 12{,}100^{\left(\frac{[-1233+T]*T}{9900+70*T}\right)} \quad (5.1)$$

While a mathematical determination of the correct concentrations would be difficult for our blood mimicking fluid given the complex nature of the multiple components, the proper percentage of each component could be easily modified experimentally. By conducting viscosity testing at 37 C at various concentrations of xanthan gum and glycerol in a phosphate buffer, the proper recipe for a blood mimicking fluid at physiological pH could be determined. This would also require the integration of a heating control element into the system.

4.3.4 Coloration

Adding color to the blood mimicking fluid could have two implications: greater visualization of the flow characteristics and aesthetics. Given the semi-transparent nature of the silicone tubing, adding a dye upstream of an injurable section, and then proceeding to injure the model could provide some information about how the flow is altered by induced injuries of different sizes. A dye could also help visualize how the fluid flows around an injected biomaterial in the vessel. In addition, a red dye would provide an aesthetic improvement to the cardiovascular model, giving it a look and feel that more closely resembles the human anatomy. This could be potentially beneficial for students learning about the fluid dynamics of the cardiovascular system as an additional visual stimulation.

4.4 Control/Feedback Systems

4.4.1 Electronics

We would eventually like to have a completely automated platform. This means that the flow data, pressure data, and flow rate through the system is measured, recorded, and controlled without any additional recording or effort on the researcher's end. The sensors could be placed and calibrated in the system and after the researcher were to run the program the test would run and the data would automatically be controlled.

A fully automated system is possible to achieve with an arduino board and the subsequent meters

[38] Nian-Sheng Cheng, "Formula for the Viscosity of a Glycerol-Water Mixture," *Industrial and Chemical Engineering Chemistry Research*, 47, 9, 2008. 3285-3288.

and sensors that are compatible with the sensor, but given more funding and with a much larger budget we would like to make changes to the whole electronic system running and measuring our data. The programming code for the arduino system reads and runs in a very linear fashion.

This means that when trying to run multiple measuring systems and do many tasks simultaneously, the arduino system slows down and some glitches may occur. In order to solve this problem, we would like to use a higher order and more powerful processing power arduino board or use an separate arduino board for each of the measuring and control systems. There would be and arduino board running the program for the flow rate, pressure readings, and the motor control.

4.4.2 Measuring Equipment

The measuring equipment was functional but in order to collect precise and accurate measurements more sensitive meters must be used. There are no other better arduino compatible flow meters on the market, but instead of using an arduino system or compatible meter we would be able to purchase and afford a much more precise and accurate meter. Although this data collection may not be automated, the better data is more than worth the trouble.

The flow meter that we used was at the least, precise within 10% of the actual reading. There are however liquid flow meters that are much more precise, just much more expensive and is not automated. An example of a much better meter is the SM Mag Flow meter. This meter can measure flow rates of up to 130 gallons in an hour, much better and much more than what is needed for our system.

In addition to a better flow meter, we would also like to change the type of pressure sensor that we used. Instead of adapting a force resistor to the purposes of our platform we would like to get a sensor that is actually meant to specifically measure liquid pressure. There are no higher or more powerful sensors that are also compatible to run on the arduino system. If we had the resources and funding a separate and more accurate meter would be purchased. As with the control of our liquid flow meter, the measuring platform would not be able to integrate this either, but only for the most top clients and special invitation.

4.5 Solutions and Solution Testing

4.5.1 Determining Workable Properties

The next step in the solution development process will be to identify what the properties of the silicone window sealant, the sealant we found effective at stopping hemorrhage, are. This will require a variety of different of different tests and evaluations. We are primarily interested in the density, viscosity, hydrophobicity, and bulk properties of the material. It is possible that some of

what we require can be obtained from the manufacturer, but it is unlikely they will release their recipe for any reason.

Further testing on the abilities of the sealant to stop simulated hemorrhage must also be conducted. Tests on buried injuries, as discussed above, would be of paramount importance moving forward. It will also be important to analyze different brands and forms of silicone sealant to identify which demonstrates the best option and the best properties.

Financial/Market Considerations

5.1 Budget and Financing

5.1.1 Initial Budget Proposal

Our original budget for the project was vastly overestimated due to the ambitious project scope we initially proposed (See Table 5.1). This budget took into account the cost for the cardiovascular system model, the blood mimicking fluid, real blood samples, and injectable material candidates. We received a maximum grant from the Santa Clara University School of Engineering for $1500 from this budget proposal for purchasing of materials, equipment, and instruments. After conducting a thorough market analysis and literature search, we eventually narrowed our focus to just the cardiovascular system inclusive of the BMF, tubing, and electronics, which lowered our cost significantly.

Table 5.1 Anticipated Budget and Financial Need
This was the proposed budget that we submitted to the Bioengineering Department at Santa Clara University.

Material	**Anticipated Cost**
Cardiovascular System Model	
Pulsatile Pump	$2,000.00
Silicone Tubing	$500.00
Dynamic Sensors	$500.00
Flow Resistors	$200.00
Miscellaneous	$200.00
Injectable Material & Apparatus	
Biomaterial Candidates	$500.00
Syringes/Injection Hardware	$300.00
Glassware	$500.00
Chemicals & Reagents	
Polyethylene Glycol	$200.00
Blood Simulation Reagents	$500.00
Miscellaneous	$500.00

Miscellaneous	
Milling/Construction Expenses	$200.00
Whole Blood Analysis	$200.00
Whole Blood Samples	$200.00
Subtotal:	$6,500.00
Taxes and Shipping Expenses (9%)	$500.00
Emergency Expense Buffer (9%)	$500.00
Grand Total	$7,500.00

5.1.2 Total Project Cost

After restructuring our budget and conducting research to spend our funding effectively, we were able to easily fit our project costs into our budget. The final cost of our entire project, inclusive of the device construction, materials, construction tools, and equipment came out to be $1477.52. The actual cost of our project can be seen in the spending report in Table 5.2 below.

Table 5.2 Total Project Cost
Some of the costs listed on this master budget are approximated but the grand total is accurate. Along the course of our project several receipts were lost or given to the Engineering Department for reimbursement. All estimates are reasonable based on actual purchases made and fair market prices.

Qty	**Material**	**Cost**
Pump and Pump Systen Components		
1	Douglas Fir 2x6x8	$4.44
9	Waxman 2" rubber rigid caster	$35.73
2	9" Springform pans	$31.98
1	Black and Decker cordless power drill 12V	$39.69
2	Milled HDPE blocks	$13.16
2	12" springform pans	$31.20
Subtotal		$156.20

Vasculature Model		
80 Feet	Siicone tubing, various sizes	$111.60
128	Polypropylene and PVC tubing fittings	$101.21
25 Feet	PCV tubing, various sizes	$17.44
2	Large plastic buckets	$5.96
Subtotal		$236.21

Blood Substitute

1	Glycerol 10kg (W252506-10KG-K)	$87.00
1	Xanthan Gum 100g (G1253-100G)	$64.10
Subtotal		$151.10

Electronics and Measurement Systems

1	Starter pack for Arduino (Arduino Uno R3)	$64.95
2	Liquid flow meter - 1/2" threaded	$19.90
2	Square force - sensitive resistor	$15.90
2	Round force - sensitive resistor	$14.00
1	Breadboard	$10.00
Subtotal		$124.75

Solutions Testing Materials

2	Loctite Epoxy	$10.64
2	JB Weld ClearWeld Epoxy	$13.04
2	DOW Insulation Foam Spray	$8.11
2	OSI Quad Window and Door Sealant	$12.16
1	Caulk Gun	$7.06
Subtotal		$51.01

Tools, Hardware, and Accessories

1	Dewalt Nutdriver Set	$38.00
2	Aluminum Billet Scraps	$4.35
Var.	Miscellaneous fasteners, washers, and ties	$50.00
Var.	Additional tools and construction supplies	$75.00
Var.	Assorted drill bits	$25.00
1	Roll of Duct Tape	$4.00
1	Partial purchase of rotational rheometer	$500.00
Subtotal		**$696.35**

Master Subtotal	**$1,415.62**
*Tax and Shipping**	$61.90
Grand Total	**$1,477.52**

5.1.3 Actual Device Cost

The cost of the actual device was much less than that total project cost, and was very competitive compared to other cardiovascular models on the market with similar properties. Looking at the

cost of the materials used only for our device from the spending report in table 5.2, the total cost for the cardiovascular model was around $583.92. This cost could be significantly reduced if materials were to be bought in bulk as well.

Table 5.3 Actual Cost of Cardiovascular Simulating Device
Costs associated with overhead or other purchases have been removed from this list leaving only costs which were incurred in the process of actually building the final prototype.

Qty	Material	Cost
Pump and Pump Systen Components		
1	Milled HDPE blocks	$6.58
4	Waxman 2" rubber rigid caster	$15.88
1	9" Springform pans	$15.99
1	Black and Decker cordless power drill 12V	$39.69
1	Aluminum Billet Scraps	$2.18
Var.	Miscellaneous fasteners, washers, and ties	$50.00
1	0.25” Plywood Board	$10.00
Subtotal		$140.32
Vasculature Model		
50 Feet	Siicone tubing, various sizes	$49.30
65	Polypropylene and PVC tubing fittings	$53.54
2	Large plastic buckets	$5.96
Subtotal		$108.80
Blood Substitute		
1	Glycerol 10kg (W252506-10KG-K)	$87.00
1	Xanthan Gum 100g (G1253-100G)	$64.10
Subtotal		$151.10
Electronics and Measurement Systems		
1	Starter pack for Arduino (Arduino Uno R3)	$64.95
2	Liquid flow meter - 1/2" threaded	$19.90
2	Round force - sensitive resistor	$14.00
1	Breadboard	$10.00
Subtotal		$108.85
Master Subtotal		**$509.07**
*Tax and Shipping**		$74.86
Grand Total		**$583.92**

5.2 Marketability

5.2.1 Market Competition

The primary competition for our was from three pulsatile pumps used in high-end cardiovascular research labs. These three models are the ViVitro SuperPump, the BDC Pumps PD-1100, and the Harvard Apparatus 55-3305, which are all shown in Figure 5.1 below. These three pumps provide pulsatile flow at flow rates and frequencies that mimic the human heart, like what our model does. These prices are for the pump alone, without any additional vasculature assembly, blood mimicking fluid, or integrated electronic sensors.

These pumps are marketed toward well-funded research groups doing novel and specific studies of benchtop cardiovascular flow. In order to complete the system, a vessel system or tissue phantom is necessary, which would significantly increase the cost. Our model is a complete system, and while not as specific as these, with more verification testing, could be just as viable a model.

Figure 5.1 Other Heart Simulating Pumps
(A) The ViVitro SuperPump, an extremely expensive heart-simulating pump that can accept computerized waveforms. (B) the BDC Pumps PD-1100, another extremely expensive cardiac simulator with a high range of adjustability. (C) The relatively simple and uncomplicated Hardvard Apparatus 55-3305, another cardiac simulator.

5.2.2 Research Market

The first market that our project has potential for is less well-funded research labs and start-up companies looking for a means to innovate in the field of traumatic hemorrhage, without the significant investment required to purchase one of the more high-end cardiovascular pumps. Our device offers an inexpensive alternative for proof-of-concept testing and initial material testing at a lower risk than the more expensive instruments. The modular design of our cardiovascular

model (pulsatlile pump, blood-mimicking fluid, tubing system, and electronic controls) could also be appealing to companies who want to customize the cardiovascular model to fit their research needs.

5.2.3 Education Market

A second major market that this pump could be appealing to is the education market. The pump is inexpensive, modular, modifiable, and potentially programmable which would offer great opportunity for elementary, high school, or undergraduate education on the pulsatile flow patterns of the cardiovascular system, the effects of traumatic injury on the flow through vessels, and how potential traumatic bleeding solutions can be applied to vessels.

With some additional electronics integration, quantitative data on flow rates and pressure changes could help students understand dynamic cardiovascular characteristics. This data could also be used to teach fluid dynamics properties such as laminar vs turbulent flow, reynolds number, wormsley number, and dimensional analysis. It is both visual and quantitative, and could become an integrated device to be used for hands-on learning of the cardiovascular system.

An additional educational opportunity would be to offer the entire cardiovascular model as a buildable kit with instructions. Students could have the opportunity to build the pump, vasculature assembly, and circuit by themselves in a classroom setting. This could allow students to learn basic skills about fabrication, electronics, and construction. For schools and Universities who offer students access to a laser cutter and 3D printer, the students could even fabricate their own personalized housing and motor mount using modifiable plans provided to them in the kit, which further increases the education value of this model.

5.2.4 Manufacturability

In addition to having viable markets in the research and educational realms, the model itself is easily manufacturable and scalable. We were able to fabricate the device using a laser cutter, 3D printer, Bridgeport milling machine, and lathe. All the parts for the assembly are easily machinable and scalable, and were designed with simplicity, functionality, and manufacturability in mind. The parts of the housing, pump, and tubing easily interface with one another to produce an assembly that can be dismantled or put together simply. All materials used are readily available to do not require specialized purchases because we wanted to make sure the model was frugal and easily constructible.

Professional Issues and Constraints

6.1 Ethical Constraints

6.1.1 Introduction

The objective of this project is to change the way we treat hemorrhagic injuries in the field, and to save lives. On the surface, this is a pure pursuit. We have an uncomplicated goal – to improve a necessary medical process – and an innocent, positive motivation – to save lives. But, at the risk of sounding pessimistic, no project is that uncomplicated and no pursuit is totally pure. Throughout history we have seen numerous efforts at "saving lives" end in disaster, and numerous projects aimed at increasing medical efficiency exploit humanity itself. Our supposedly simple project is necessarily and positively complicated by a variety of ethical concerns. To do our due diligence as engineers, designers, and stewards of the Earth (in true Jesuit fashion) we must address these ethical concerns and adopt appropriate standards up front. We must additionally do our best to stay abreast of additional concerns as they arise. With this maxim in mind, we have identified the following major areas of concern as regards the ethics of our project. Beneath each major area can be found the standards we have decided to adopt to address that area.

6.1.2 Outline of Ethical Principles Adopted

I. Team Interactions, Integrity, and Moral Codes
    a. Standard of Interaction
    b. Standard of Integrity
    c. Standard of Morality
II. Social and Societal Impact and Usability
    a. Standard of Societal Involvement
    b. Standard of Access
    c. Standard of Function and Reliability
III. Product and Design Ethics
    a. Standard of Ethical Reporting
    b. Standard of Complete Design
    c. Standard of Safety

6.1.3 Team Interactions, Integrity, and Moral Codes

6.1.3.1 Standard of Interaction

Almost all modern ethics encourage an active concern for justice. Typically we think of justice in large terms – the fairness of the law, etc. – but notions of justice pervade everything we do. Our interactions as a team should be just. No one person should be doing an unreasonable percentage of the work, nor should any one person be operating in a vacuum. Our team will adopt a standard of fair interaction premised on our commitment to treat each other fairly, divide work evenly, criticize constructively, and make decisions with the good of the whole in mind. This ethical standard also, more than others, shapes all of our approaches to those below. An ethics of fairness and justice in our interactions will ensure proper checks and balances on any one team member straying from the standards below.

6.1.3.2 Standard of Integrity

It makes no sense to commit to an ethic or standard of fair interaction without also committing to an ethic of integrity. If all of the members of our group are treating each other equally, but no one is maintaining any form of integrity, nothing we do is valid. It is a duty implicit to our aims, as beneficiaries of generosity of this University, its faculty, its staff, and its students that we maintain academic and personal integrity. It is a duty implicit to our aims that we truthfully pursue success. The only ethical alternative is to relinquish our right to pursue this project and to return our funding in full. This will be demanded of us and will be something we demand of ourselves if we reach a point in our work beyond which we cannot proceed with integrity.

6.1.3.3 Standard of Morality

The final standard we must describe and adopt before moving on to the ethics of the project itself and its outcomes concerns our general attitude to right and wrong. It is irresponsible and incompatible with our previous standards to ignore the existence of right and wrong approaches in all that we do. We cannot work collectively in a vacuum any more than any one of us can without jeopardizing the value, justice, and ethics of our project. It is our intent, then, according to our pure pursuit, to make all choices pursuant to this project in full knowledge of the possible consequences, having considered the individual righteousness of each, and having selected the most ostensibly righteous. It is our intent further, as mentioned previously, to adapt our approaches if the apparent morality of a choice or decision changes. We will not make choices calibrated only to achieve positive ends for our project or persons. In summary, we adopt a standard of making all decisions with full knowledge and full consent of a morally inclined will.

6.1.4 Social and Societal Impact and Usability

6.1.4.1 Standard of Societal Involvement

There are two tightropes this project must walk related to its interactions with society. The first concerns how society will interact with it. An ethical respect for self-determination, termed autonomy, increasingly drives medical decision making. As prospective makers of a medical

device we must be open to many perspectives and open to taking risks to bring great good into the world while remaining conscious of autonomy and its implications regarding misuse. We must walk the line between making a product that is effective and the best it can be, and making a product that is hard to abuse. This can also be interpreted by way of the principles of beneficence and non-maleficence. We must design something that will not ultimately lead to harm if used correctly and minimizes the harm if used incorrectly (non-maleficence). It is not enough, though, to leave it there – we must also ensure that it accomplishes some good for the user if used correctly (beneficence). The second tightrope concerns how much society can know about the design. We must walk the line between providing true and accurate information, and keeping some of the details and functions of our device secret (to ensure healthy competition, long-term feasibility, and preventing alarmism).

6.1.4.2 Standard of Access and Frugality

Following from our standards of societal involvement are our standards of access. There is an ethical duty, in medicine, to respect life in all forms and work to maintain it, save it, and improve it where possible. This imperative applies regardless of any financial gain to be gained by the treater or any characteristic of the treated. In our position, we would be pseudo-caregivers: the device our treatment, capable of saving lives. Thus, we have an imperative to prioritize the saving of lives and caring for of patients above any material gain so associated. This requires us to seek, above all else, to improve the quality of life for the world by allowing more people to develop life-saving treatments. If we are to truly seek this goal, we need to ensure that our device can be purchased and operated extremely frugally, and we need to ensure that it is highly accessible to those willing to do research and make a difference. At the same time, though, we need to make sure our device is not accessed by those who do not have any intention of pursuing safe and well-motivated research.

6.1.4.3 Standard of Function and Reliability

Also following from our standard of societal involvement and our standard of integrity is our standard of function and reliability. As the developers of a medical device for use in testing in extreme conditions, we have an ethical imperative to ensure that our design will live up to the expectations of the entire range of the target audience's expectations. If we provide individuals with a device that we say does X, and they observe that the device does not do X, we have done them harm (maleficence) and failed our ethical obligation not to deceive and to be of aid to those in crisis. Furthermore, if our device not only fails to do X, but does Y instead, we have failed our ethical obligation to provide our autonomous users full information regarding our product. We have robbed them of their right to informed consent. Consequently, we have a duty to provide a device that functions as indicated and continues to function as indicated under the indicated circumstances. Perhaps this is merely a specific derivative of earlier standards; if it is, that is good – these particular thoughts are of such crucial importance to our ethical design they can hardly be overstated or over-mentioned.

6.1.5 Product and Design Ethics

6.1.5.1 Standard of Ethical Reporting

As a corollary to our standard of academic integrity above, we find it relevant and useful to identify a standard of ethical reporting. It is crucial, as discussed ad nauseam heretofore, to be honest in our reporting of what our device can and cannot do. To do so accurately and ethically demands an attention to even the finest details of data collection and analysis. Care will be taken in tests to achieve statistical randomization in selection, assignment, and treatment throughout any studies. At no point will we knowingly misreport any data, figures, analysis, or results. We will be extremely careful, with an eye to the moral and ethical implications of failing to do so, to avoid any accidental misreporting as well.

6.1.5.2 Standard of Complete Design

No design exists outside of a context. Our project's proposed model of the cardiovascular system is only one instrument of many required for the testing of new arterial hemorrhage treatments. We must be sure that our design can be seamlessly integrated into the medical research and development process. The design must be complete, and must show respect for the complete process of medical solution development. Care must be taken to ensure that our device can actually substitute for testing on other, more expensive and involved, physiological models.

6.1.5.3 Standard of Safety

This is perhaps the most self-explanatory of the many standards we have outlined here. In order that all involved be treated justly, and that no one be harmed, we establish the safety of those designing, manufacturing, distributing, and using our proposed solutions to be the primary focus of our pursuits. There will be no exceptions to this rule of safety. If things become unsafe, they will be shut down and abandoned. Everyone will do his/her part to ensure that we are functioning as a safe team.

6.1.6 Conclusion

We will make ethical perspectives a central part of our research, design, and development discourse as we move forward with our project. We will be particularly guided by the 10 ethical standards we have outline here and the general ethical principles of autonomy, beneficence, non-maleficence, and informed consent.

6.2 Aesthetic Constraints

6.2.1 Introduction

The goal of this project is to change the way that we, as a global society, treat and manage traumatic hemorrhage when we are outside of the hospital or clinic setting. We seek to redefine innovation in traumatic hemorrhage by allowing more people into this field. This project, then, stands to benefit emergency medicine, military medicine, and developing world medicine equally. Not one of these medical demographics has traditionally been concerned with classical aesthetics: choosing instead to adopt the mantra, “If it works, who cares how it looks.” Yet, aesthetics has to do with a lot more than making things look pretty; it has to do with balancing their function with their form in context, and under this definition has much to do with our project. The form of our proposed device ultimately must call to mind its function, and vice versa. Our device may not look pretty, in the traditional sense, but it will have to look and feel simple, professional, and reliable: things that cannot be communicated by the function alone. It is important to additionally note the importance of the aesthetics of communication regarding our project.

6.2.2 Analysis

6.2.2.1 Simplicity

The concept of simplicity can be interpreted as having two distinct subcategories relevant to our project: functional simplicity – i.e. usability, and formal simplicity – i.e. minimalism. The first facet is important to our project because of our audience and market demographics. The kind of people this device will be of use to (military and civilian medics, developing world health care providers/persons) do not have time to waste “figuring out” a device or reading complicated instructions. They will not have consistent training, language skills, or a solid understanding of human anatomy/physiology. The device must be designed so that its function is explicitly clear from its form and that its operation requires very few steps. Any instructions necessary should be printed directly on the device in pictorial form when possible. Equally important is the device’s formal simplicity. The environments it may be used in as a testing platform will by necessity be rugged, dirty, and chaotic. Any excess stylization will provide an opportunity for avoidable contamination of the product, trip-ups and malfunctions upon operation. Clean, uncomplicated lines should prevail. Professional, but informative color schemes should be employed.

6.2.2.2 Reliability

While less tangible than simplicity, the appearance of reliability is no less important in our design. The people using this testing platform have to trust that it will simulate the human body accurately, and that it will work better than other options at their disposal. A device which looks unreliable, for all intents and purposes *is unreliable* as far as a potential client is concerned. Because this is a medical device for the testing of new traumatic hemorrhage solutions, a reliable

appearance requires it to be both sturdy and serious looking. As a result, the polymers used in the construction of the design should be sturdy plastics in substantial thicknesses. The device should be designed to withstand significant impact and shear forces. All joints should be sealed and sterile components protected by at least two layers of membrane/hard material covering. Nothing on the device should be easily caught or snagged, and nothing should easily break off. Unlike medical devices for the general consumer, resources should not be wasted on excessively ergonomic or sculpted features that would both add weight and trivialize the device. Attention to device aesthetics in this case requires a commitment to not making things more attractive or beautiful than they should be.

6.2.2.3 Professionalism

An aesthetic of professionalism implies all of the following: 1) a commitment to the balance between quality and frugality that allows for good, but not wasteful product design; 2) design that is respectful to the financial and environmental needs of its user; 3) a professional approach to the design that accounts not only for the aesthetic of the design itself, but also of the process and the context of the design itself. Things designed to high standards of quality by professionals have a certain feel and appearance, by definition a certain aesthetic. This device must carry the same "feel." The user must be able to detect in the device the care for the user's purpose that went into its design.

An aesthetic of professionalism also demands a commitment not to waste or over-engineer. Professionally made devices aren't excessively flashy, ornate, or overly intricate. Our design must be frugal enough to be broadly applicable in developing world medicine. This means it must be as inexpensive as possible, but no cheaper. Excess will not be entertained in design, but no necessary feature or characteristic will be omitted or overlooked. This will carry into the design of the device and blend slightly with the aesthetic of simplicity above. It is also imperative that the design process be aesthetically sensible, simple, and complete. The approach we will take will be modeled on the iterative approach encouraged in engineering.

6.2.2.5 Literature/Presentation

Aesthetics in communication are more easily understood than the more abstract design aesthetics that pertain to other areas of our project. Perhaps the most crucial forms of communication in engineering include the graphical and data-oriented. These are closely followed by the verbal forms of communication we are dealing with almost exclusively in English 181. Our graphical communication will be created to be legible for a wide variety of readers – per our audience analysis.

We will leave the highly specific and technical details for written appendices and figure descriptions, because the figures themselves should merely communicate the basic concepts and general flow. Pleasing and sensible colors will be used (blue for deoxygenated blood flow, red for oxygenated blood flow as is convention, etc.). Simple shapes and flow diagrams will also be employed (circular diagrams for circuit/iterative processes, arrows for flow direction). Care will be taken to edit all documentation and presentations, and to practice extensively for any verbal communication attempts.

6.2.3 Conclusion

Our design is not intended for use by the common consumer. Our product is not designed for sale in ordinary markets, but rather fits better into the government and humanitarian contract markets. As such, the traditional aesthetic requirements that consumer designs must accommodate do not apply to the same degree. Nevertheless, many aesthetic concerns apply to our design. It still must be taken seriously and chosen above others by its intended audience. It must look and feel trustworthy and well-made. It must be well presented and described in all forms of communication. We must remember these aesthetic concerns if we are to change society's approach to the treatment of traumatic hemorrhage.

Summary and Conclusion

7.1 Project Accomplishments

7.1.1 Project Accomplishments Overview

Addressing the issue of innovations in the field of traumatic hemorrhage is a huge undertaking. Innovations in this field require a multidisciplinary approach, involving fluid dynamics, human physiology and anatomy, material science, thermodynamics, mathematics, and chemistry. Our original intent was to develop a usable solution for treating all types of traumatic hemorrhage, including both internal and external, arterial and venous systems. We soon found that this task was impossible if we didn't have a viable testing platform of the human cardiovascular system, so we set out to develop one.

We were successful in this endeavor, for we have developed a viable, pulsatile model of the cardiovascular system that pumps a non-newtonian blood-mimicking fluid through a system of branching silicone tubing that can be injured using a fabricated peristaltic pump. There is still verification, automation, and optimization testing needed, but we have developed a strong and functional foundation for future work on this project and in this field.

7.1.2 Pulsatile Peristaltic Pump

The pulsatile peristaltic pump design that we currently and employing was inspired by a different pump design and adapted to our purposes. Our pump is a frugal solution to mimic the flow rate of the human heart. Most other pumps available on the market to mimic the parameters of the heart are more than 10 fold the cost of the materials for our pump design. The pump design works by creating a vacuum type suction in the tubing by having a wheel occlude the tubes against the side of our pump housing, then pulling fluid through with the lead wheel and pushing the fluid through with the trail wheel. This type of pumping action creates the pulsatile flow by only forcing a set amount of fluid through the tubing at a time.

7.1.3 Tubing System

We designed the tubing system based off of the blood vessels and vasculature of the human body, specifically focusing on replicating the diameter of the arteries as closely as possible. We decided that we should focus on testing the artery areas because injuries in the field to a patient's arteries is where the greatest risk of death is, when an artery is damaged the patient can easily bleed out and suffer from hypovolemic shock. We used straight and Y couplings to set up our system, stepping the diameter of the tubing down in size at each level to replicate the body's

vasculature of arteries turning into arterioles and eventually into capillaries. The step down in diameter went from ½" tubing all the way down to ⅛" tubing. For the purposes of testing material solutions this change in size was more than enough.

7.1.4 Blood-Mimicking Fluid

We have developed a novel blood-mimicking fluid that mimics the physical properties of human blood like nothing we found on the market or in our extensive literature review. Our fluid not only mimics the viscous and non-newtonian shear-thinning properties of blood (other research have done that) but it also mimics the pH of blood that no other research has showed so far. Additionally, our blood-mimicking fluid has the potential to easily be adjusted to match physiological salt concentrations and physiological pH, allowing for greater versatility of testing solutions. This fluid has many implications for projects outside of this one as a physical blood-mimicking fluid for benchtop physiological flow testing where real biological blood is not necessary or would over-complicate the system.

7.1.5 Integrated System (talk about electronic data collection here)

Our project was initially meant to be controlled completely by an arduino board. We used an Arduino Uno board as well as a arduino compatible flow meter and force sensitive resistor to control and collect quantitative data for our system. We used the data to prove numerically that the design of our pump met the same parameters as the human heart. Although with our initial design and data collection the pump was run by an electric drill and the flow was controlled by the trigger of that drill, in our final design we plan to use a modified motor of a battery operated drill. The motor speed and therefore flow rate would be controlled through the arduino using a technique called Pulse Width Modulation, sending rapid signals of turning off and on the drill to increase or decrease the rpm of the drill head.

7.2 Project Impact

7.2.1 Impact of project on Society

The current state of our project has very real implications on our society in multiple fields. The primary impact our project has on society is the introduction of a frugal but fully functional effective cardiovascular model that can be injured, of which there is nothing exactly like on the market or in research. Our project has the potential to create its own market and its own research opportunities with its low cost and unique functionality specifically for traumatic bleeding simulation. It dramatically lowers the cost of entry new players to innovate and experiment in the field of traumatic hemorrhage, opening the field to new ideas and new opportunities that once were only available to well-funded laboratories that could afford the expensive pulsatile pumps that are the competition to this project.

Given the new opportunities available from this platform, this project could lead to new solutions to traumatic hemorrhage that could save lives on the battlefield or in civilian traumatic accidents, as well as potentially lead to the creation of new start-up companies that could have a positive economic impact in this field.

7.2.2 Future Implications

While the current state of the cardiovascular model already has real societal implications, with just a few modification to the model, the reach of the impact could be expanded further. Once the pump control and data collection are fully automated, it would increase the efficiency and repeatability of testing with this model, making it competitive with even the high end pulsatile pump systems on the market. With this cardiovascular model could have an impact on education from the elementary to college level by allowing students to actual assemble the device themselves, build the circuit, look at fluid flow through the system, and try their own bleeding solutions. It could also enter into new research areas beyond traumatic hemorrhage.

Due to the modular characteristic of the device, the pulsatile pump could be separated from the tubing and be used for ultrasound flow measurements to look at flow profiles and non-newtonian flow in a pulsatile flow phantom system. The pump could also be used with real biological blood and blood vessels: the injurable section could potentially be replaced with real arterial tissue to get a more accurate representation of the cardiovascular system. The reach of this project is wide, and the potential is great. Once the system is optimized, this pulsatile cardiovascular platform could be a viable option for innovation not only in the field of traumatic hemorrhage, but also in many other fields that could have use for a model that mimics physiological conditions.

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