Medical Simulation Testbed for Interventional Cardiology and Neurology

A Major Qualifying Project

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Abstract

Medical testbed systems are used to mimic various physiological conditions of the vascular system. There are current designs for medical testbed systems that can help researchers to experiment with new interventional methods and physicians to practice procedures that will treat cardiovascular diseases. However, there is a need to economically design a tissue-mimicking and anatomically accurate human artery simulator. This project aims to design, fabricate, and validate an artery simulator that can represent calcified atherosclerotic lesions and simulate different pressure waveforms of blood flow within the body. The final design consists of an artery model made of clear polyethylene terephthalate (PET) which connects to a diaphragm and piston pump in series, separated by a back-pressure regulator and check valve. Design verification methods showed that the system was able to successfully achieve an anatomically accurate structure and pressure waveform of blood flow.

Acknowledgements

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Executive Summary

Cardiovascular diseases are the leading cause of death for men and women in the United States, and can be caused by factors such as (but not limited to) medical conditions, lifestyle choices, or genetics [1]. A major component of the cardiovascular system includes arteries which are used to transport nutrients and oxygen to organs in the body. Composed of smooth muscle, these vessels dilate and constrict to create an arterial pressure waveform consisting of a systolic and diastolic pressure [2]. As healthy blood flow is characterized by having a laminar quality and endothelial shear stress, disruption in flow can lead to a variety of cardiovascular diseases [3]. A current, although imperfect, method of treatment includes the insertion of a stent into affected blood vessels. Stents act as short-term solutions as they often need to be replaced due to failure from wear or damage of local tissue.

To explore improvements and new methods for intervention of cardiovascular diseases and help device development and medical training in interventional procedures, we propose the use of a medical testbed that can be designed and constructed. In order to meet this need, our project aims to design a system and medical testbed that accurately fabricates the physiological structure and pressure waveform of blood flow in the cardiovascular system within a reasonable budget.

Four broad objectives of the device were defined: reliability, reproducibility, biomimicry, design integration and ease-of-use. These objectives were separated into sub-objectives that further describe the attributes of each broad objective. From these objectives, design constraints related to performance, cost, resources, and schedule were identified. From the design objectives and constraints, a revised client statement was synthesized, specifying the project would produce a simulator that could produce a pressure waveform between 60 and 200 mmHg in the internal carotid artery.

Based on literature review and direct comparative analysis of existing arterial testbeds, the team designed and assembled the testbed that can be seen in Fig. 1. The schematic consists of a constant flow rate diaphragm pump that establishes the mean pressure; a pneumatic cylinder in parallel with a linear actuator that is modulated by a stepper motor to create the pressure wave amplitude; a back pressure valve to negate reverse liquid flow; an arterial system plastic model; a pressure transducer to provide pressure readings at a peripheral location; and finally a peripheral relief valve that adds a further mode of pressure modulation. The pressure transducer and stepper motor are both controlled by Arduino Uno modules. A feedback loop was created between the pressure transducer readings and the linear actuator to calibrate the amplitude of the pressure wave in real time.

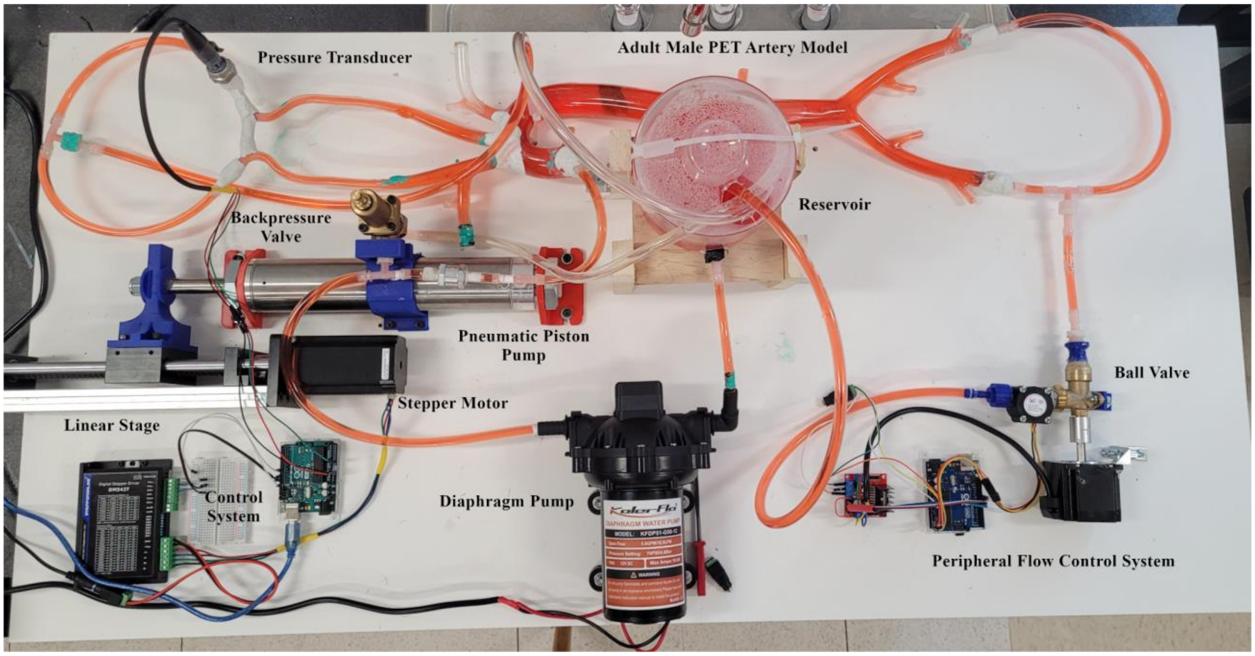

Figure 1. Assembled Medical Testbed

For statistical analysis, the data from the pressure transducer was stored and exported to MATLAB to create a graph of the pressure wave for validation that the system met the functional objectives.

Fig. 2 demonstrates the output readings of the pressure transducer over several cycles of the system. As can be seen, the system was able to modulate between 120 mmHg and 80 mm Hg based on the feedback response of the set pressure maximum and minimum values. The generated waveform had a mean offset of 0.989 mmHg and a normalized RMS error value of 1.441 mmHg.

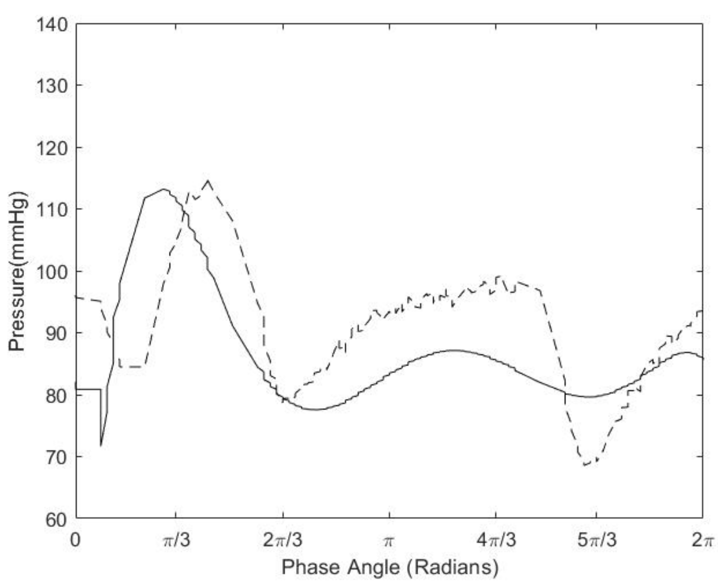

Figure 2. Output Pressure Waveform at Internal Carotid Artery

The simulator was able to produce a pressure waveform according to the design constraints. The mechanical and electrical components performed reliably. This project produced a pressure waveform comparable to the human circulatory system as well as expensive devices discussed in the literature review, with far more affordable components.

In the future, incorporating a GUI to toggle between different diastolic and systolic pressures could be developed to simulate different physiological conditions. Additionally, the pressure waveform could be set to a set to follow a more complex model than a sine wave to reduce the error. This device should also be tested with a blood-mimicking fluid to better replicate the hemodynamics of blood vessels. Lastly, developing an arterial vessel system that better mimics the compliance and viscoelasticity of blood vessels in vivo would improve the biomimicry of the system.

Chapter 1. Introduction

Cardiovascular diseases are the leading cause of death worldwide [1]. A major component of the cardiovascular system includes arteries, which are used to transport nutrients and oxygen to organs in the body. Due to being surrounded by smooth muscle, these vessels dilate and constrict to create an arterial pressure waveform consisting of a systolic and diastolic pressure [2]. As healthy blood flow is characterized by having a laminar quality and endothelial shear stress, disruption in flow can lead to a variety of cardiovascular diseases [3]. Current treatment methods include the insertion of a stent into affected blood vessels. Stents act as short-term solutions as they often need to be replaced due to failure from wear or damage to local tissue. There is a need for interventional treatments for cardiovascular diseases with better success rates and life spans.

Medical testbeds can be used to simulate physiological conditions such as the pulsatile arterial blood pressure waveform. Researchers use medical testbeds to understand the physiological conditions of the human arterial system in order to better treat these conditions by understanding the pathology of different cardiovascular diseases. Physicians and surgeons use medical testbeds to practice surgical procedures to ensure that they have the proper training necessary to complete these procedures in the environment of an operating room. Although helpful in the healthcare and medical research field, medical testbeds can be very expensive with high-priced parts that may be unaffordable to researchers or physicians in resource-limited areas. Laboratories or hospitals in these areas may not have access to the resources or funds to purchase testbeds, creating the need for a cost-efficient and affordable system.

To increase the accessibility to medical testbeds and the ability to explore improvements and new methods for intervention of cardiovascular diseases and help device development, we propose designing and constructing a medical testbed with inexpensive and reliable components.

There are several approaches for designing medical testbeds to simulate different physiological conditions depending on the goals of the researchers. Designs can either focus on a small section of the cardiovascular system, such as a single artery, model the flow patterns but not true geometry of the cardiovascular system, or use a full artery model. There are also different approaches to generation of flow and pressure waveforms such as syringe pumps, gear or diaphragm pumps, and piston pumps. These pumping systems can either be controlled utilizing variable direction or speed motors, or controlled flow valves.

The pumping system presented in this report was developed to generate accurate pulsatile pressures at the internal carotid artery utilizing a full abdominal cardiovascular model within a reasonable budget.

Chapter 2. Background

2.1 Relevance of Medical Testbeds in Clinical Applications

Ischemic heart disease and ischemic stroke are the two most prominent cardiovascular diseases worldwide [4]. The treatment of these diseases could improve greatly based on the learnings that can be taken away from a mechanical medical testbed system. All cardiovascular diseases relate in some way to the obstruction of healthy and mainly laminar blood flow in blood vessels. Endothelial dysfunction is defined as a “pathological condition characterized by an unbalance between vasodilatory and vasoconstrictory mechanisms,” and is considered to be an early development leading to most cardiovascular diseases [3]. Healthy blood flow is classified as being laminar and having a high endothelial shear stress (ESS). On the other hand, unhealthy blood flow is turbulent and low ESS, therefore, disrupting endothelial Nitric Oxide (NO) production. Understanding these diseases and researching possible improvements to the current standard treatments is an extremely prevalent issue in healthcare, as the number of cardiovascular disease deaths rose from 12.1 million in 1990 to 18.6 million in 2019 [4]. In addition, a mode of intervention that is very commonly used today, inserting a stent into the blood vessel, is not considered to be a long term fix to the problem because a larger percentage of patients require secondary intervention due to stent failure. Stents can fail mainly because they alter the hemodynamic forces and stresses on local blood vessel tissue.

A medical testbed that can accurately and reproducibly mimic various physiological conditions of the vascular system is a powerful tool for experimenting with new means of intervention for cardiovascular diseases before implanting them in the body. This research could be used to improve the long term capabilities of stents or even for developing tissue engineered artery grafts that have the potential to replace stents as the standard intervention method. Researchers found that engineered grafts oftentimes are not capable of withstanding the stresses of pulsatile blood flow and degrade until eventual failure [5]. Therefore, it is necessary to simulate the likely stresses that in vivo blood flow will put on a graft in vitro to avoid graft failure and further intervention.

Beyond tissue engineering, the medical testbed could be used as a controlled environment for surgeons to practice their skills on specific cardiovascular procedures that aim to reestablish healthy arterial blood flow including atherectomy and thrombectomy. Ultimately, the goal of the project is to use the medical testbed in conjunction with a camera system. Combined, these two tools will realistically simulate the environment of an operating room such that surgeons can learn and finetune skills.

2.2 Artery Physiology

Arteries are a major component of the circulatory system that consist of tubelike structures to transport nutrients and oxygen to organs in the body [2]. They are composed of smooth muscle to constrict or dilate, creating an arterial pressure waveform within the body that results from catecholamines being released into the blood [2]. Systolic pressure is formed when the heart contracts, while diastolic pressure is formed when the heart is at rest between contractions as the heart expands and refills with blood. There are approximately 20 arteries in the human body that carry oxygenated blood - with the exception of the pulmonary artery - throughout the circulatory system. A simplified diagram of the human arterial system can be seen in Fig. 3.

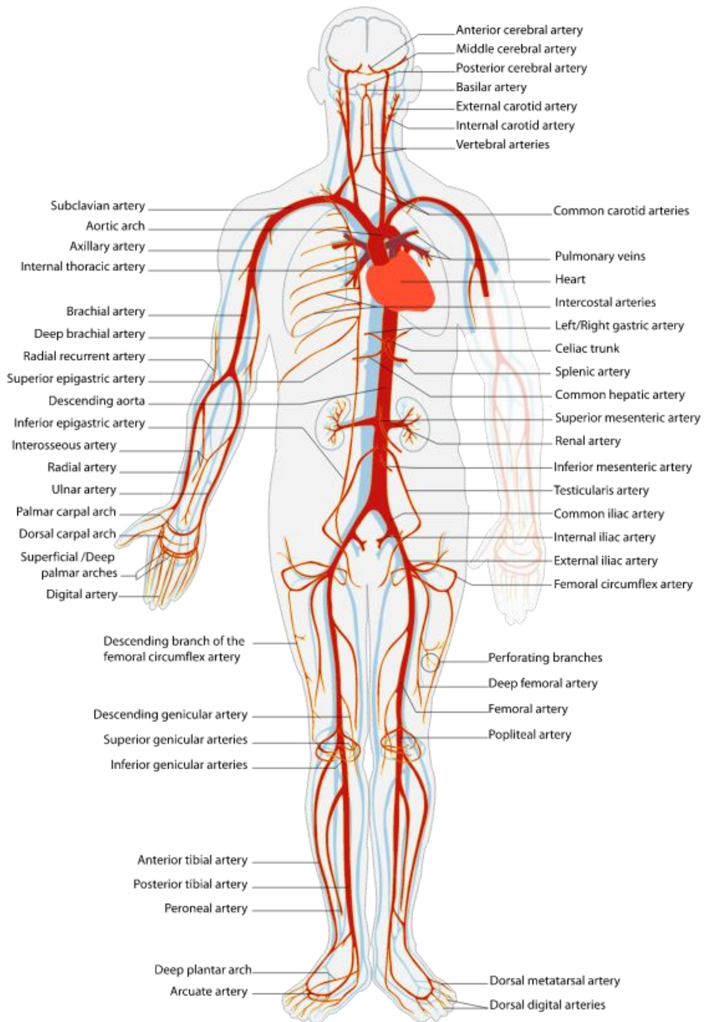

Figure 3. Simplified diagram of the human arterial system [6].

To effectively replicate an arterial model, it is important to understand the physiology of arteries and their flow and function. Arteries have three layers where the first innermost layer is the intima, the tunica is the second innermost layer, the second layer is the media, and the outermost layer is the adventitia [7]. These layers are depicted in Fig. 4.

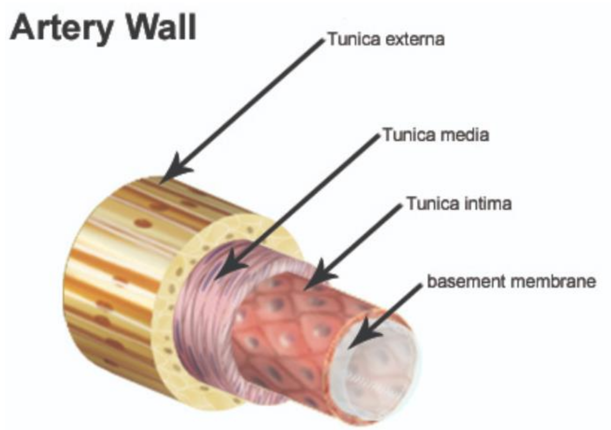

Figure 4. Diagram of artery wall with its three layers [7].

The varying compositions of each arterial layer allow for arteries to work to transport blood throughout the body and contract, creating different pressures and assuring the blood reaches the most peripheral capillaries of the body.

Arterial blood flow starts in the pulmonary artery which leads blood to the lungs, while all the other arteries move blood to different parts of the body. Blood is pumped from the left ventricle through the aorta which splits into four sections that each carry it to different areas in the body [7].

Finally, understanding the flow rates and pressure that can be seen in blood vessels is essential for best replicating in vivo conditions. Uematsu et al., (1983) found that blood flow rates in the body reach their fastest values between 5 and 20 years of age with a mean of 8.5 ± 1.3 cc/sec [8]. To understand the most extreme pressures that could be seen in the body, a study on subjects with a history of circulatory diseases such as hypertension or ischemic heart disease was conducted. The study found the peak systolic blood pressure in these patients to be 150 mm Hg [9].

2.3 Carotid Artery Pressure Waveform

An artery pressure's waveform is composed of the steady and pulsatile components of the artery's pressure. The systolic pressure of the heart is arterial pressure during heart contraction while diastolic pressure is the minimum arterial pressure between contractions when the heart expands [6]. Many studies – some of which are described in section 2.4 – have been conducted using a variety of methods to determine the most physiologically accurate waveform of an average human adult. With these values, one can effectively replicate the waveform through different models and designs.

There are many different methods to determine a carotid artery pressure waveform as documented in literature. The gold standard and earliest form of assessing an arterial pressure waveform is an invasive technique that requires the cannulation of a peripheral artery [10]. Although it is able to yield accurate results, this technique is used in limited studies as it is not preferred for the comfort of test subjects in studies. Thus, many non-invasive methods for measuring an arterial pressure waveform have been created. Applanation tonometry methods are able to determine both central and peripheral waveforms of carotid or radial arteries, and these waveforms can then be modeled by using mathematical functions [10]. Although non-invasive, this method is challenging to use, inaccurate, and ineffective, requiring specialized equipment. Two modern and non-invasive methods include echo tracking and Tissue Doppler imaging (TDI). These methods are more common with advancements in technology and can also provide insight into lumen diameter and assessment of degree of stenosis for atherosclerosis diagnosis [10].

TDI is the method commonly practiced in research, and various studies have been able to determine the pressure waveform from this method. From using ultrasound waves to show blood moving through blood arteries and veins, colour doppler imaging can determine the velocity information of blood flow [11]. Using this method, one study was able to determine the internal diameter waveform and profile of blood flow velocity of the right common carotid artery as seen in Fig. 5.

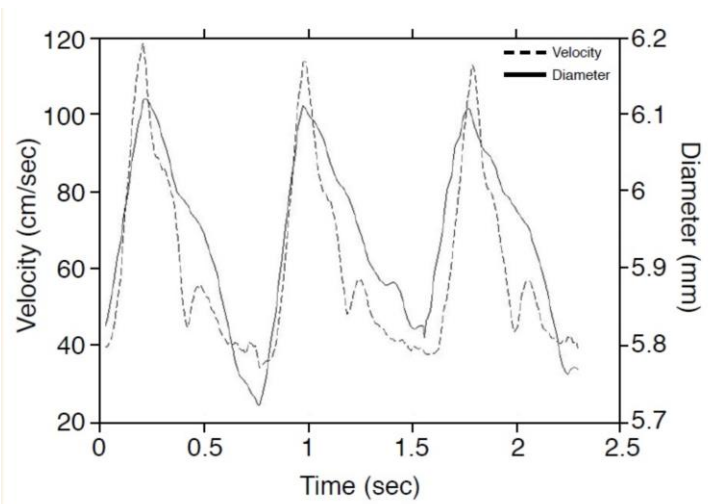

Figure 5. Internal diameter waveform (solid line) and profile of blood flow velocity

With this extracted velocity, the internal diameter waveform could be determined. Thus, sonograms in this study were able to determine the pulse pressure waveform of the common carotid artery, which is depicted in Fig. 6.

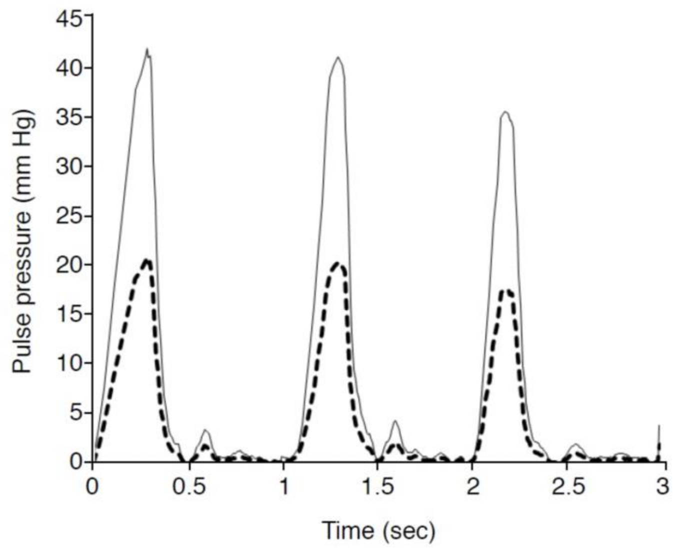

Figure 6. Pulse pressure waveforms of the common carotid artery of a healthy subject [10].

In accurately developing a pressure waveform that replicates that of the body, one can replicate the blood flow of an artery in a model to be physiologically accurate. Determining the arterial pressure waveform is essential for ensuring a system reflects the blood flow of the human arteries, independent of velocity profiles.

2.4 Arterial Pressure Simulator Testbeds

There are a wide variety of peer reviewed papers that describe successful design and replication of pulsatile arterial blood pressure waveforms. In addition, the means by which different researchers have achieved this also vary. The approach and design of different testbeds vary depending on the goals of the researchers and the arterial areas of focus.

2.4.1 Comprehensive Endovascular Test Bed

Reddy et. al. [12] developed a versatile endovascular testbed designed for simulating arterial conditions for mechanical thrombectomy for strokes consisting of a syringe pump, variable flow resistors and latex tubing. They accurately generated physiological pressures with pulsatile waveforms to simulate healthy patients (120/80mmHg) and patients who have suffered stroke (147/85mmHg) [12]. This design utilized two different stages of arterial models: a general cardiovascular model, and a cerebral artery phantom. The general system layout can be seen in Fig. 7.

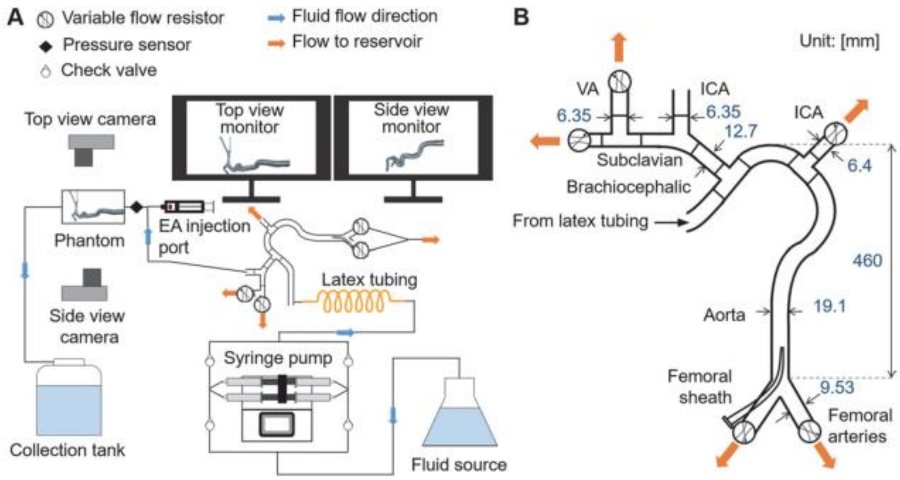

Figure 7. Testbed schematic with general cardiovascular model geometries [12].

The general cardiovascular model replicates the paths and diameters of the cardiovascular system, but not the true geometry as it is outside the area of focus.

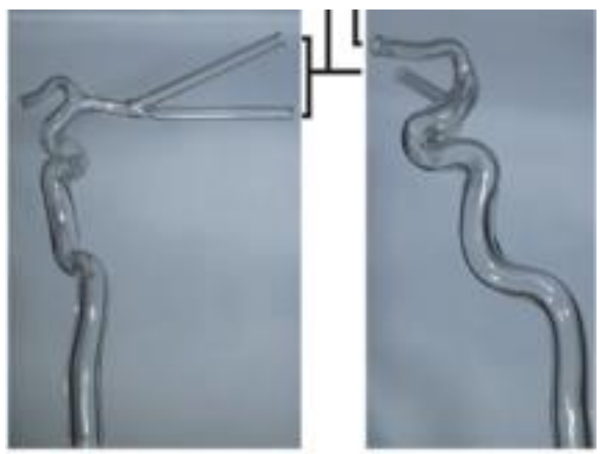

Figure 8. Silicone Cerebrovascular Phantom [12].

The cerebral artery phantom seen in Fig. 8 was a custom designed model which utilized patient CT scan imaging to create a 3D CAD model which could be manufactured out of plastic, glass, or silicone [12]. The design developed by Reddy et. al. [12] shows a good balance between modeling actual vasculature in the area of interest, biomimetic flow patterns, and a simple pumping system.

2.4.2 Novel Perfusion Bioreactor

Nandan et al. [13] was one of the first papers that used a multi-channel peristaltic pump to replicate pulsatile blood pressure in series with a media chamber and a pseudo vessel. The authors used this to create a 3D in vitro bioreactor model for analysis of stent impact on endothelial cell (EC) health. They produced continuous pressures of 120/80 mmHg with a pulse frequency of 1 Hz. In addition, it was the first time the results were able to confirm results on EC health post-stent deployment after more than 24 hours. The results showed disturbances to hemodynamics and injury to local EC all caused by the stent. The system schematic can be seen in Fig. 9.

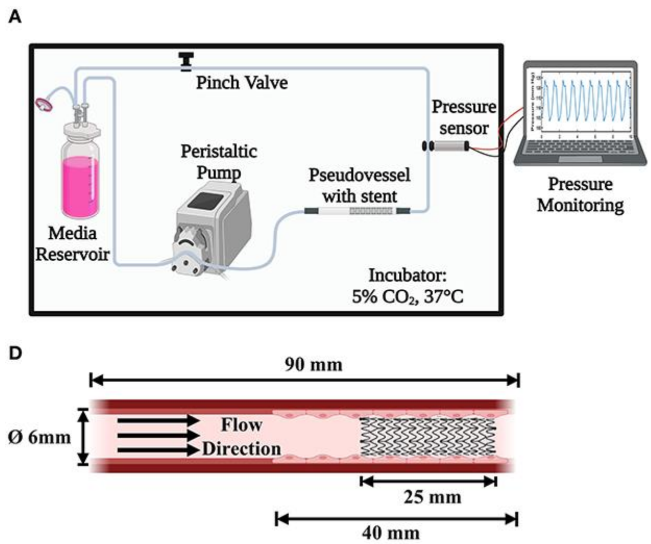

Figure 9. Peristaltic Pump Bioreactor System [14].

With its focus on the pseudo vessel and biomimetic artery walls, the pump design in Nandan et al. [13] is simple with a single flow path powered by a peristaltic pump. While this system is straightforward, commercially available peristaltic pumps cost upwards of $5000 [14].

2.4.3 Radial Pulsation Simulator

Yang et al. [15] developed a two-stage radial pulse simulator using a linearly actuating piston that simulates the left ventricle systolic pressure wave and a pressure control reservoir that models diastolic pressure [15]. The system models the abdominal cardiovascular system with a radial artery model in parallel diverging at the subclavian artery [15]. In addition, the model includes an artificial silicone thoracic aorta to alter arterial stiffness similar to what happens naturally with age. They were able to successfully replicate the pressure waveform of the radial artery, including the early systolic pressure, late systolic pressure, and the dicrotic notch. A diagram of the system can be seen in Fig. 10.

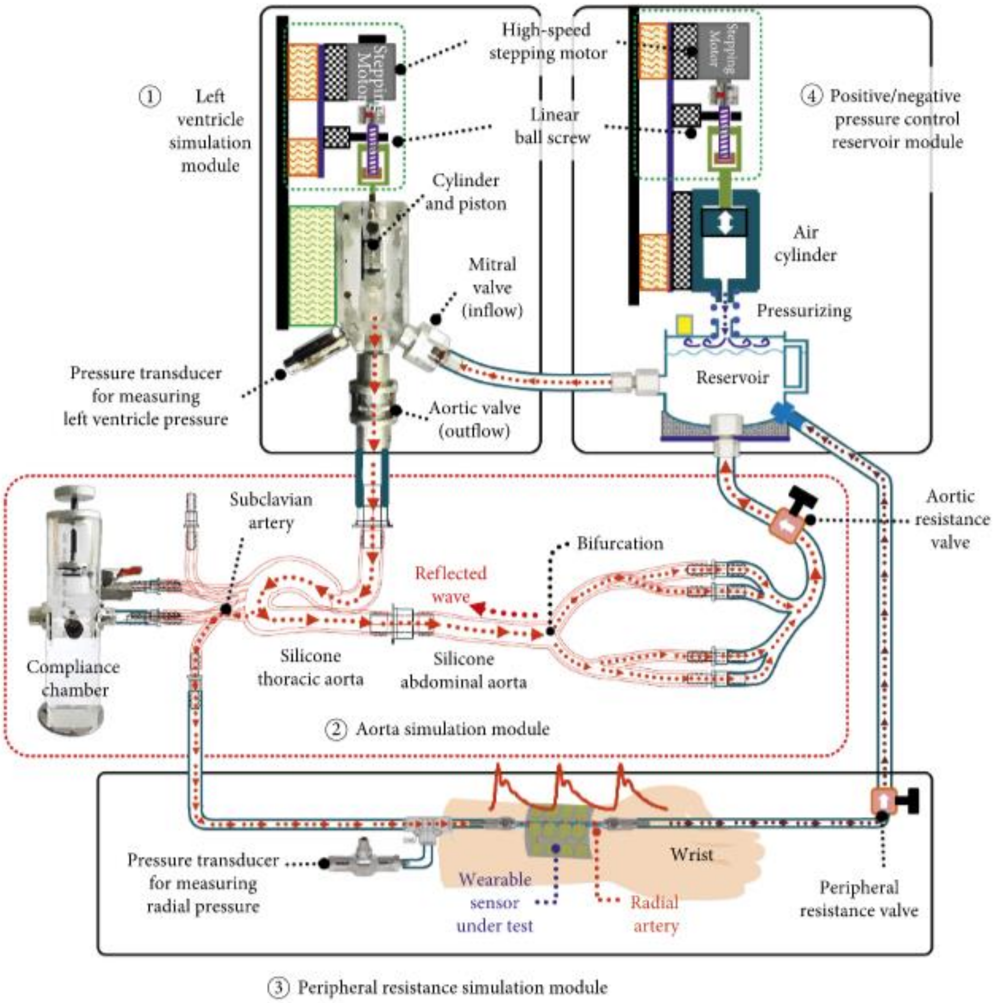

Figure 10. Radial Pulsation Simulation Testbed [15].

Yang et. al. [15] provides a complex take on a medical testbed model through both its use of a full artery model, peripheral focus area, and biomimetic two-stage pumping system.

2.4.4 Flow pumping system for physiological waveforms

Tsai and Savaş [16] designed a simpler two-stage pumping system utilizing a gear pump in series with a linearly actuating piston pump simulating similar but simpler systolic/diastolic stages [15]. A schematic of the system is depicted in Fig. 11. The system does not focus on replicating any particular vasculature with shape or material characteristics. Instead it focuses on the reliable generation of sinusoidal, coronary, and carotid waveforms [16].

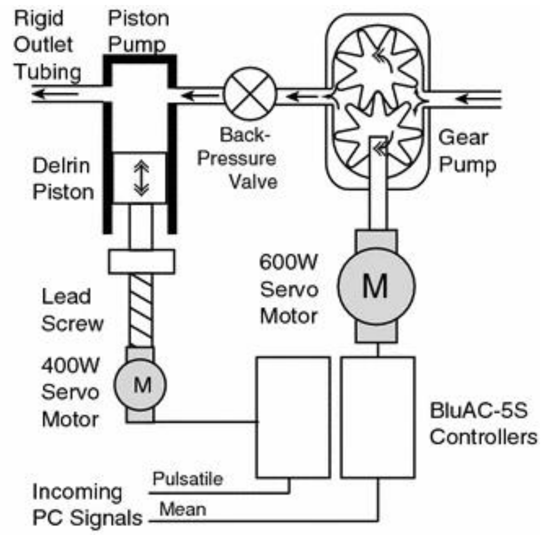

Figure 11. Two-Stage Physiological Waveform Testbed [16].

2.5 Control Systems

As defined by the National Institute of Standards and Technology, a control system is a system in which deliberate guidance or manipulation is used to achieve a prescribed value for a variable [17]. Generally speaking, there are two types of control systems, open loop and closed loop. In a basic system, a control system is defined by the following elements: input, controller, and output. The input being the provided signal, the controller being the internal or external element of the system that controls the desired process by modifying or amplifying the inputted signal, and the output being the overall response of the system achieved after the controller's modification of the input signal [18]. In an open loop system, an input signal (or command) is applied, amplified in a controller, and its output received in an output element. This type of control system exists when the output of the system depends on the input, but the input (or controller) is independent of the output of the system. This type of system does not contain any feedback loops and can thus also be called a non-feedback system [18].

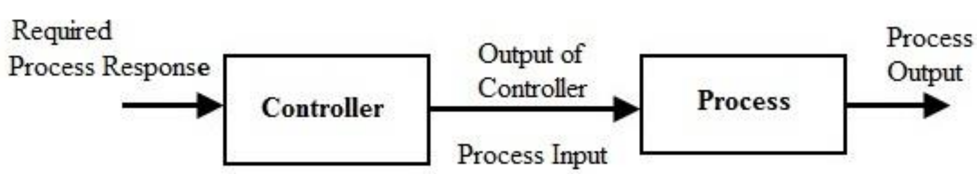

Figure 12. Simple Open Loop Control System [18].

Now, in a simple closed loop system, the controller is no longer dependent on the input, but instead by an error value. This error value is defined as being the difference between the system input and its output. A closed loop system has the same basic elements found in an open loop system, except it contains two additional features, an error detector and a feedback loop. The error detector is the component used to determine the difference between the desired output and the actual output and then produces a signal proportional to the difference between the input and output signal. Additionally, the feedback loop is the component in a system in which some proportion value based on the system's output is used as an input for future operations [18].

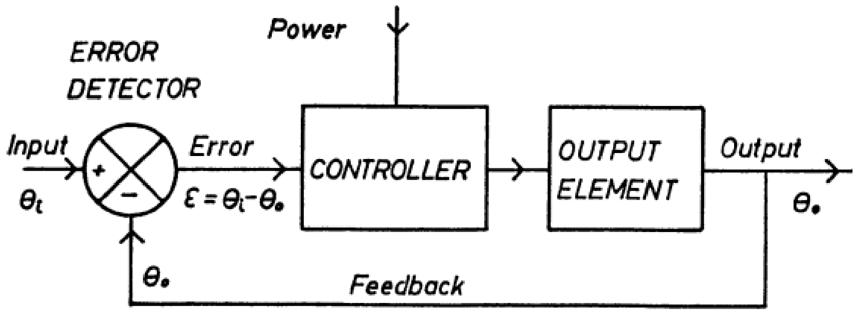

Figure 13. Simple Closed Loop Control System Examples [18].

As a continuation of the open loop and closed loop systems, they can be further categorized as either: a feedback loop or feedforward loop. In a feedback loop, the system continuously measures the output and compares it to the desired setpoint, while making adjustments to the input until the output reaches the desired setpoint. This type of control loop is used in applications where the output is difficult to predict and is found in common devices such as thermostats, where temperature is measured and adjusted to a desired set point [19]. In feed forward loops, the system adjusts the input based on external factors known to affect the output, without relying on measurements of the output. This type of control loop is used in applications where the input can be adjusted to compensate for external factors during the process. Common examples are found in applications such as aircraft control where the system adjusts engine power based on changes of altitude or airspeed [19].

As a widely used application where both feedback and feed forward loops can be combined to provide accurate and reliable control over a wide range of processes/systems, a Proportional, Integral, and Derivative (PID) Control system is employed. PID Control loops combine both feedback and feedforward loops to adjust a process to a setpoint. The proportional component of the PID Control loop provides immediate adjustments to the input based on the current error between the setpoint and the output. The integral component accumulates the error over time and provides a corrective factor that increases as the error is still present. The derivative component provides an adjustment based on the rate of change of the error which helps prevent overshooting or instability in the system [20]. All three components when working together allows for the PID control loop to provide stable and precise control over a range of differing applications. The three main equations are as follows:

Proportional:

$$X = K_C(x_{set} - x) * t + c$$

Integral:

$$X = \frac{1}{\tau}\int_{0}^{t}(x_{set} - x)(t)dt$$

Derivative:

$$X = \tau_d \frac{d}{dt}(x_{set} - x)(t)$$

The proportional, integral, and derivative elements of the PID loop can be adjusted by the user until the system is behaving as desired. Altogether, the three terms are added together to form the main equation in the PID loop and is as follows:

$$u(t) = K_p\left(e(t) + \frac{1}{T_i}\int_0^t e(\tau)d\tau + T_d\frac{d}{dt}e(t)\right)$$

In the terms of a medical testbed, a PID Control loop can be used to adjust the pressure toward a desired value depending on the phase of the waveform. Extreme values, or maximum and minimum pressures, can be set that will inform the system when to oscillate back in the other direction. Additionally, a desired waveform can be implemented so that the system is trying to follow a set path between the extreme values.

Chapter 3. Project Strategy

3.1 Initial Client Statement

The initial client statement serves as an identification for need and as a basis for further design discussions between the design team and the client. Dr. Yihao Zheng provided the team with the following initial client statement:

> *"Design, develop, and validate a electrical and mechanical simulator which can be used to reproducibly and cyclically generate a cyclic pressure waveform."*

3.2 Defining the Stakeholders

As mentioned, the client, Dr. Yihao Zheng, detailed the expectations for the completion of this project to the design team. Further guidance on detailed needs and wants of the project is the client's primary role. This will be done to refine the project as the design team moves towards the final product in order to ensure this aligns with the client's vision. Additionally, the end users, the surgeons and physicians who will be practicing on the medical testbed, should be considered to ensure that the final product will be useful. Either direct feedback or consultation through Dr. Yihao Zheng will be critical to evaluate day-to-day and end user functionality. It is the goal of the design team to integrate the needs, wants, and any additional feedback from the client and the user to develop a product that fits the client's vision. The initial client statement will evolve with additional, more detailed and specific, discussions of the needs of the client and user.

3.3 Developed Objectives and Design Constraints

Based on background research and further discussions with the client, a list of objectives was created to guide the development of the system. These objectives inform design tradeoff decisions by being ordered by importance by the client, end user, and design team. The broad objectives established were *reliability*, *reproducibility*, *biomimicry*, *design integration, and ease-of-use*. These broad objectives were then broken down further into sub-objectives. Table I below outlines the objectives and their associated sub-objectives. Table II gives further specificity to how the design team defined each objective and sub-objective as they relate to this project.

Table 1. System Functional Requirements

Objective	Description
Reliability	
Durability	The device should be able to undergo an experiment without experiencing failure or significant wear. The fluid flowing through the device should not corrode or damage the mechanical and electrical components.
Encased Design	The design should primarily be self contained to avoid damage during usage or transportation.
Reproducibility	The output of the device should demonstrate precision and accuracy, so results of a test or experiment are easily reproducible.
Accurate	The device should be able to output pressure within 5% to what the control was set to.
Precise	The pressure outputs of the device should be measurable within a specified range for each cycle.
Biomimicry	
Pressure Waveform	The device should generate a physiologically accurate pressure waveform that can match provided systolic/diastolic pressures, pulse shapes, and pulse rates.
Flow Characteristics	The device should have flow characteristics that mimic physiological conditions (i.e. limited air suspension, limited backflow).
Material Characteristics	The arterial model should have the compliance of an artery while mimicking the material of a vascular graft.
Design Integration	
Compact Design	The device should be compact to allow for integration into the full medical testbed.
Ease-of-Use	

Easy to Monitor	A GUI aids a user to achieve the settings they desire without needing knowledge about the inner workings of the device.
Easy to Program	The device should provide a simple way to program the pressure waveform to allow for simulation of different conditions.

Design constraints are criteria that the project must absolutely meet to be determined as successful. These are determined by both the client, end user, design team, and the university. These design constraints are outlined in Table II below:

Table 2. System Design Parameters

Constraint	**Description**
Must be able to create a cyclic pressure waveform	- Primary client goal
Pressure must be measured at a location representative of a human artery	- Primary client goal
System must be appropriate for pressures ranging from 60 mmHg to 200 mmHg	- Modeling human physiological conditions
Must cost less than $3000	- Budget provided by WPI for a team of five students with additional funding from client
Must be achievable by April 2023	- Must be completed before WPI's project deadline and graduation

3.4 Revised Client Statement

Utilizing the design objectives and constraints detailed above, the design team reconceptualized the initial client statement. The following is the revised client statement:

> *"Design, develop, and validate a electrical and mechanical simulator which can be used to reproducibly and cyclically generate a clinically accurate pressure waveform between 60 and 200 mmHg in the internal carotid artery. The system should be able to run continuously for one hour and not accumulate significant errors or wear during use. The system should be compact to allow for integration with a camera rig system."*

Chapter 4. Design

4.1 Design Process

Since arterial blood flow simulations have been done many times in the past, it stood to reason to begin the design process by conducting a literature review. This serves to get an understanding of what has been done successfully in the past and where there is room for improvement in terms of performance and cost. The team's literature review shown in section 2.4 was concentrated around locating papers that had successfully met the parameters, functions, and specifications of the client statement. From there, the group used a Pugh Matrix to directly compare how different alternatives stacked up based on their performance in terms of the project's function and specification requirements as well as the total cost. From the results of the Pugh Matrix, the team created an initial design concept and began to order parts that met the specifications necessary to mimic the characteristics of the human cardiovascular system. Once the system was assembled, the various control systems and pressure sensors were coded, calibrated, and adjusted. Once the general blood flow pressure wave had been observed, the team ordered new parts that would allow for the system to reach higher pressures, and therefore, better mimic the human cardiovascular system.

4.1.1 Analysis of Design Ideas

In order to determine the best possible system for our project, the team used the Pugh Matrix method. This method helped our group outline the criteria necessary for our project, and objectives for the final design of our project. The Pugh Matrix table can be seen in Fig. 14:

Objectives	**Cost**	**Reliability**	**Reproducibility**	**Biomimicry**	**Design Integration**	**Ease-of-Use**	**Total Score**	**Weight**
Cost	X	1/2	1/2	1	1	1	5	0.3125
Reliability	1/2	X	1	1	1	1	4.5	0.28125
Reproducibility	1/2	0	X	1/2	1	1	3	0.1875
Biomimicry	0	0	1/2	X	1	1	2.5	0.15625
Design Integration	0	0	0	0	X	0	0	0
Ease-of-Use	0	0	0	0	1	X	1	0.0625

Figure 14. Design Parameter Pugh Matrix Table

The six main objectives that originate from the client statement were implemented in the matrix. Thus, the final deliverable should be cost-efficient, reliable, able to produce precise and accurate results, able to mimic the physiological conditions of our targeted area, able to be integrated easily into the overall testbed, and should have straightforward instructions that are easy to understand. The group used the Pugh Matrix to directly compare each objective and determine which ranked higher in terms of necessity. This would serve as a guide during system design when questions of which components should be purchased before others. The objective each component satisfied and the rank of said objective over others would settle these debates.

The group then ranked each system from Chapter 2.4 in terms of each objective and calculated a weighted rank score. The Pugh Matrix Table for our concept selection can be seen in Fig. 15. Based on the calculations done, the optimal design to use for our project would be the gear/piston pump, as this concept is able to meet most of our design parameters.

	Weight	**Baseline**	**Cylinder and Piston System**	**Gear/Piston Pump**	**Cam-based Simulator**	**Progressing Cavity Pump System**
Cost	0.3125	0	3	4	3	2
Reliability	0.28125	0	3	4	3	4
Reproducibility	0.1875	0	4	3	4	4
Biomimicry	0.15625	0	4	4	4	3
Design Integration	0	0	1	2	1	1
Ease-of-Use	0.0625	0	1	3	2	3
Rank Score			3.21875	4.375	3.28125	3.15625

Figure 15. Concept Selection Pugh Matrix Table

4.1.2 Initial Design

Based on the results of from the Pugh Matrix, the team created an initial design schematic, seen in Fig. 16, that loosely replicates the design of the referenced gear/piston pump system referenced in the literature review [16] while advancing the complexity by including an artery model and peripheral flow.

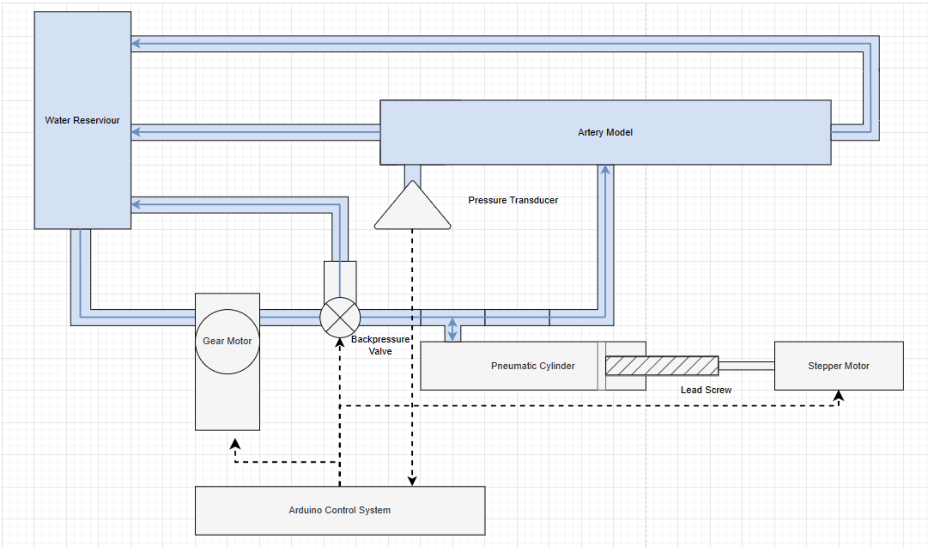

Figure 16. Diagram of Initial Design

4.2 Final Design

The finalized design assembly can be seen in Figure 17. The components and and their respective functionalities are further explained in the subsequent sections of this chapter.

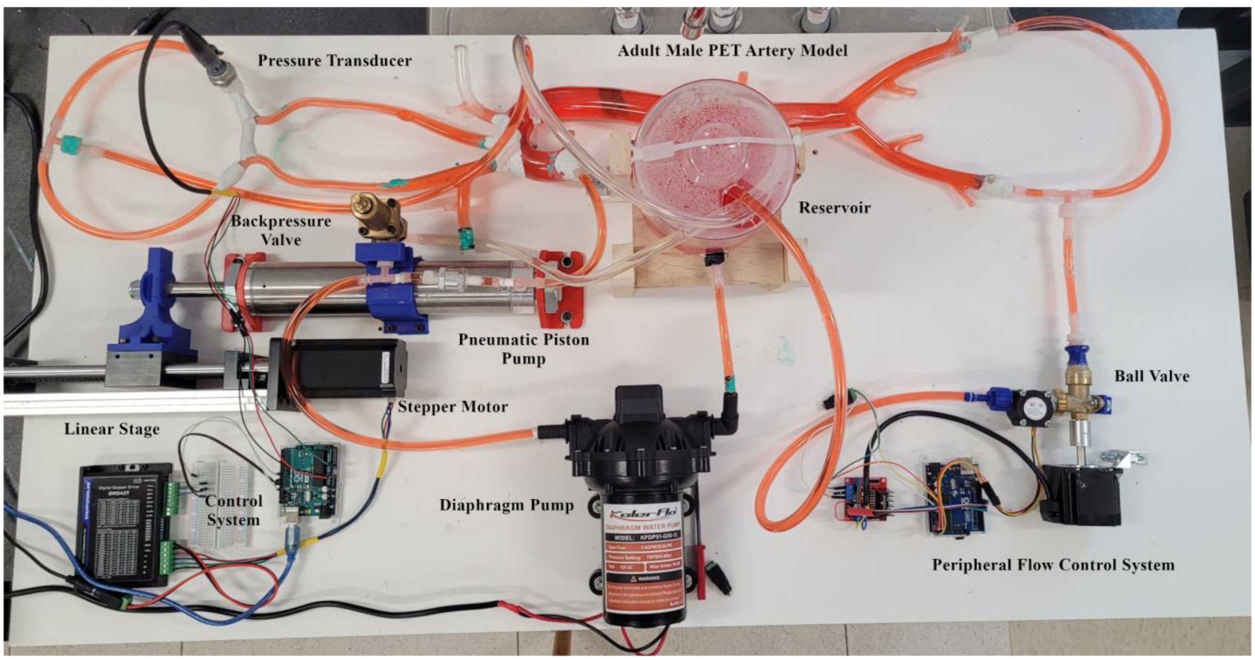

Figure 17. Final Design: Assembled Medical Testbed

4.2.1 Artery Model

The artery model used is an adult male arterial system made out of 1 mm thick clear PET. This model was chosen for its anatomically accurate geometries, low cost, and transparency for device visualization. The smooth hollow veins allow for insertion of pacemaker leads, catheters and other devices that might be needed to simulate surgical procedures.

4.2.2 Pulse Generation

The mechanical components of the pulse generation system consists of a gear and piston pump in series, separated by a back-pressure regulator and check valve. The gear pump provides a steady-state pressure component for the pulse waveform. For this design setup a KolerFlo 51 Series Industrial Water Pressure Pump (KolerFlo, USA) is used with a maximum throughput of 315 ml/s. A back-pressure valve and check valve sit at the diaphragm pump’s discharge. This setup allows for the control of the steady-state pressure of the system with excess pressure and flow vented through the valve’s exhaust port and prevents backflow into the diaphragm pump. For this design setup, a Emerson K-series Cash Valve (Emerson, St. Louis, MO, USA) is used. The diaphragm pump is run with a constant voltage to generate the mean pressure regulated by a low-level voltage trigger relay controlled by an Arduino Uno R3 (Arduino, New York, NY, USA).

The method for generating the oscillatory component of the pulse waveform is the piston pump. This system consists of a stainless-steel cylinder with a 2 inch bore and 4 inch stroke length and a 6 inch linear stage driven by a stepper motor. For this design setup, a BIMBA 314-XP-00MC Round Body Cylinder (BIMBA, University Park, IL, USA) is used. The stepper motor used to drive the piston pump is a Lin Engineering 5718C-08P-RO Stepper Motor with a rated maximum speed of 3,700 rpm (Lin Engineering, Morgan Hill, CA, USA). The motor is connected to a two-phase digital stepper motor driver (StepperOnline, Nanjing, China) which outputs step size, velocity, and directional control using an external signal. For this design setup the control signals are generated by an Arduino Uno R3 (Arduino, New York, NY, USA).

The flow pattern can be characterized as a closed loop with two branches: the carotid branch, which is the branch of primary interest, and a peripheral branch, which helps regulate mean pressure. The diaphragm pump is supplied by a reservoir which is elevated by 6 in to avoid cavitation in the inlet tubing. Rigid plastic tubing connects the left internal carotid artery to the pressure transducer to avoid corruption of the pressure waveform. The pressure at the internal carotid artery is measured by a Walfront 10 psi gauge pressure transducer (Walfront, Lewes, DE, USA). Flow through the peripheral branch is regulated by a brass ball valve and flow rate is measured by a DIGITEN G3/4" water flow hall sensor (DIGITEN, ShenZhen City, China).

4.2.3 Control System

The control aspect of the system consists of a combination of different hardware and software devices working together to produce a desired waveform. Our system consists of a pressure sensor, used to measure water pressure of the system, a stepper motor, used to increase/decrease the system’s pressure, and an Arduino Uno R3, used as the main processing unit. The Arduino microcontroller collects pressure values read from the pressure transducer, compares it to the desired waveform to be generated, and in-tandem with a Proportional, Integral, and Derivative (PID) Control Loop, calculates the rate and speed at which the stepper motor must be driven in order for the system to produce the desired waveform.

Pressure at the internal carotid artery is measured by a Walfront Pressure Transducer, capable of outputting a signal of 0.5 Volts to 4.5 Volts, which translates to a pressure reading of 0 to 500 mmHg. By means of an Arduino Uno R3 its built in 10-bit Analog-to-Digital converter, the analog voltage signal output by the pressure transducer is converted into a digital, millimeter of mercury, for easier processing. Using the Arduino programming language, a simple moving average filter is created to process the pressure transducer's readings. A moving average filter was implemented by sampling 'x' number of readings before then averaging those 'x' samples together to produce a final value. This moving average filter allows for the signal to be smoothed, preventing short term overshoots/noise, and allows for invalid data to be rejected.

Once the data is free of noise, it moves onto the PID Control Loop, which was once again programmed using the Arduino programming language. In programming the PID loop, the proportional, integral, and derivative coefficient values were tuned using the following method. The integral and derivative gains are set to '0' and the proportional coefficient is increased slightly until an oscillation is observed in the output. Following this, the integral coefficient is increased until any offset is corrected for the time scale of our system. Lastly, the derivative coefficient is increased until the system's overshot in the output is reduced to a sufficient level. Now that the PID control loop is tuned, the PID loop compares the 'actual' value (found through the pressure transducer) with the 'desired' value of the waveform attempting to be generated. Using the coefficients found in a previous step along with the error calculated between 'actual' and 'desired', the PID control loop computes the weighted sum of the three terms, and then applies a corrective value to the system until the desired output is achieved. In our case, the PID Control loop determines whether the control signal driving the stepper motor should be 'HIGH' or 'LOW' based on the corrective value produced. In other words, this decision is determined by whether the 'actual' pressure reading falls above or below the 'desired' pressure reading. Following the PID Control Loop, the Arduino outputs a signal of 'HIGH' or 'LOW' to the stepper motor driver which either drives the motor in a clockwise or counterclockwise direction, respectively. This is done in order to increase or decrease the current 'actual' pressure of the system. Over time, the system is able to reliably track and recreate any input variable waveforms.

Chapter 5. Design Verification

Various tests were conducted to evaluate the performance of the pumping system. The system's output response was defined by measuring the output pressure at the ICA as compared to a fixed waveform. To determine the system's step response, the output response to a 0.5 Hz square wave with a 50% duty cycle was measured. To determine the frequency response of the system, the output response to fixed amplitude sinusoidal waveforms with frequencies ranging from 1 to 8 Hz were measured.

Fig. 18. shows the step response of the system. The step response given the current PID controls shows a narrower pressure band than the input signal by ±7.5mmHg with oscillations ±5mmHg. The peak response is around 10mmHg higher than the input signal. This is indicative of an underdamped system. Further dampening of the system, done by increasing the derivative response, the system shows a peak phase delay of 0.040 seconds, or approximately $\frac{\pi}{6}$ radians. This delay is likely caused by time delays inherent in the computing of the PID control and communication with the stepper driver, as well as potential mechanical delays in the actuation of the piston pump. However, the time delay indicates an acceptable response rate and could be accounted for during experimentation.

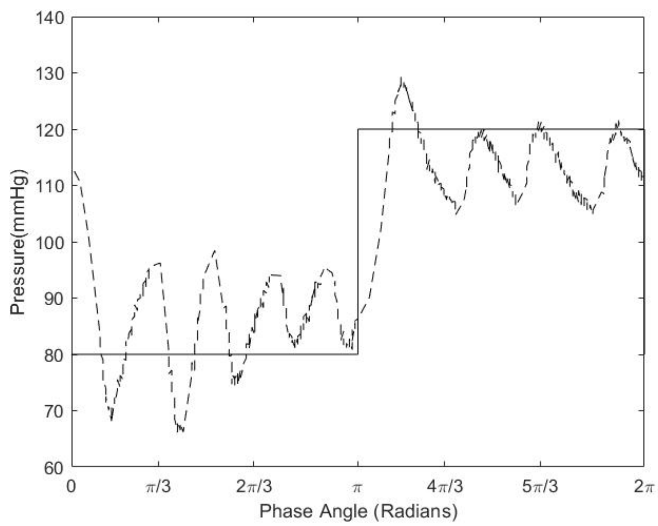

Figure 18. Step response of the system.

Fig. 19. shows the plotted results from a typical sinusoidal test performed at a frequency of 1 Hz. The mean output pressure at the ICA (dashed line) and the maximum and minimum measured extrema (dotted lines) are plotted against the sinusoidal input set pressure (solid line). At this frequency the mean output overshoots the peak by ~7.5 mmHg, creates a double hump, and undershoots the trough by ~4 mmHg.

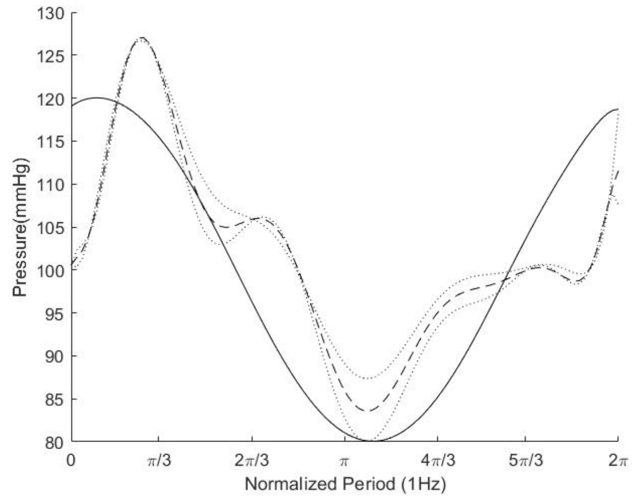

Figure 19. Comparison of the mean output flow (dashed line), spread of the extrema (dotted lines), and sinusoidal input waveform (solid line).

Fig. 20 shows the frequency response to sinusoidal inputs. There are only small magnitude response variations across the 0.5-8 Hz range, which accounts for normal physiological conditions. It is possible that with a more aggressive control system or at higher target pressures the magnitude response will decay at higher frequencies or show more variation.

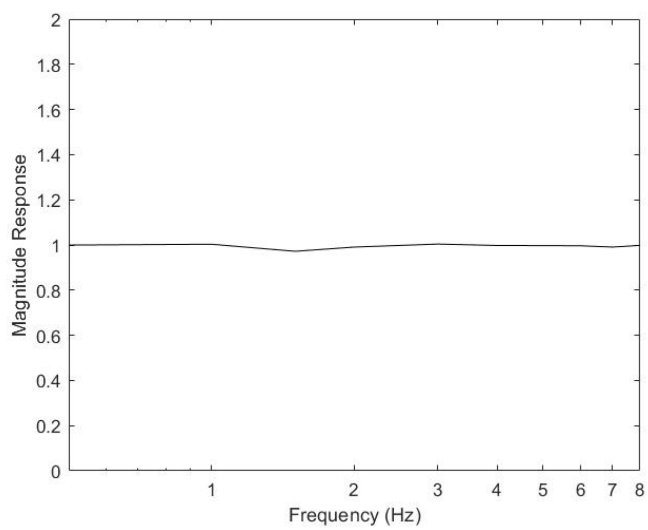

Figure 20. Frequency response of the system.

Fig. 21 shows the system response to a n=6 regression model of the ICA pressure waveform.The mean output pressure at the ICA (dashed line) is plotted against the sinusoidal input set pressure (solid line). The mean output pressure nearly matches the input waveform with only a slight phase delay and overshoots of approximately 5 - 10 mmHg.

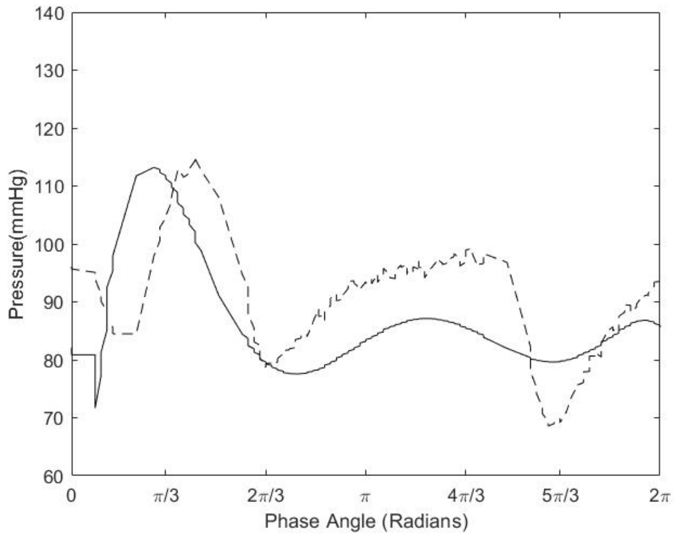

Figure 21. Comparison of the mean output flow (dashed line) and input ICA waveform regression (solid line).

Chapter 6. Discussion and Conclusions

6.1 Discussion

The difference of the average values from an entire cycle between the set pressure and the measured pressure is 0.989 mmHg, indicating the system's baseline pressure function is working well. This baseline pressure is produced by the diaphragm pump and regulated by the back-pressure valve, meaning the tuning of the back-pressure valve is sufficient. The total error of 1.44 mmHg is comparable to other pumping systems discussed in Chapter 2. This error is a result of the control system, which can be advanced with further testing and development. One way to develop the system is to tune the PID parameters on the current closed-loop control system to generate a more accurate waveform. Another possibility is that an open-loop control system might be better suited for this application. The environmental conditions do not vary significantly over time, so an open-loop control system is worth investigating in the future. A limitation of our hardware is that Arduino Uno needs time to process data, similar to any microcontroller. The microcontroller will communicate the motor to keep actuating even if the pressure has already been overshot, since there is a processing time that needs to elapse before the microcontroller knows the pressure has been overshot either too much or too little.

One difficulty of working with fluids is properly sealing each joint in the piping system. When testing new iterations of the design, the testing was slowed and needed to be delayed due to leaks between joints. This meant having to re-seal at the joints and/or 3D printing new parts that provided a better fit to avoid leaking. This delayed testing because revising 3D models, manufacturing the models, then sealing the part into the system takes time during which the system cannot be tested.

6.2 Conclusions

This project provides an economic alternative to other much more expensive medical test beds with our entire system costing less than $2000. The design uses widely available components from vendors that ship all over the world, allowing for reproducibility in most laboratory settings. The average pressure error of 0.989 mmHg shows good functionality from the first stage in the pumping system. Although the total error of 1.44 mmHg is comparable to other systems, further research and development by future project groups can improve performance while maintaining a budget-friendly design. This project can be integrated into a more complex medical testbed.

6.3 Future Work

This project was the first step of a few to developing a multicomponent system that could be used for both surgical training and intervention research applications. While this team was working to design and assemble a physiologically accurate testbed for surgical training, another team was working at the same time to develop a remote controlled camera system that would further replicate the environment of an operating room for interventional cardiology or neurology. Additionally, there are various aspects of this project that could be further developed to improve accuracy with respect to the physiology of the human body.

First, though the team was mostly successful in replicating the desired pressure wave, the curve that the system was set to follow isn't the most anatomically accurate it can be. The regression curve has the general shape of an anatomical pressure waveform but lacks the aggressive pressure generation during systole and a slow and gentle relaxation during diastole. Therefore, developing a mathematical pressure wave more accurate to that of the cardiovascular system would be a logical next step. Another limiting factor with respect to the control systems is the ability for the controllers to implement feedback when the system hits the set extreme values. As can be seen in the generated graph in Fig. _, the system would reach 125 mmHg before it was able to register and communicate to the system that it had to oscillate back toward lower pressures. Therefore, a more advanced control system than Arduino would allow for the data to be sampled quicker and respond. Additionally, implementing a GUI so that the system could be set to different physiological conditions such as hypertension with the click of a button would be beneficial for users to be able to see and measure the differences in fluid flow between a healthy and at-risk patient. Lastly, an arterial system that better replicates the viscoelasticity and compliance of blood vessels than the plastic model in this study was able to do would improve the accuracy of the system.

Chapter 7. Broader Impacts and Ethics

7.1 Project Ethics Statement

Throughout the duration of the project, the team was aware of and adhered to the Code of Ethics of Engineers to assure that our project remained aligned with the original goal to improve the state of the healthcare system and what it can do for the entirety of society.

With the project being one that could have clinical applications and direct impact on the health of human beings, it was even more important that the health and safety of society was taken into consideration. From the inception of the idea to the completion of the project, the team felt that it was working to help people by improving the way that clinicians can care for them. Additionally, the project remained within the scope of each team member's capabilities such that no harm could be done out of incompetence or lack of knowledge of possible repercussions. Furthermore, the team made sure to conduct ample amounts of literature review in preparation for the project to understand what is already known in the area of research and for guidance. The team cited the source of all knowledge or design ideas that were found to be previously done by other parties during literature review. Finally, as will be seen in the subsequent sections, the team took the time to reflect before, throughout, and at the conclusion of the project on the environmental, economic, global, and social effects of the project.

7.2 Global

Ischemic heart disease is the deadliest cardiovascular disease worldwide, accounting for approximately 16% of the world’s total deaths [21]. Since 2000, the annual deaths due to ischemic heart disease have grown from 2 million to 9 million [21], so there is a great need for an experimentation model to investigate improvement in cardiovascular intervention. When creating a model that must be suitable for populations all over the world, differences in environment and lifestyle must be considered. Many resource-limited areas in the world do not have access to experimentation models to develop new interventional techniques, given their expensive nature. This project uses sensors, actuators, and plastic parts that are widely available all around the world. The total cost for these parts is much less than medical test beds currently on the market. The team anticipates that this project will provide an alternative to help train physicians in areas of the world that face barriers in purchasing expensive medical testbeds.

7.3 Social

Although the system and artery model can mimic the pulse pressure waveform of blood flow, it does not account for the various sizes and physiological differences between male and female patients. The size of the model mimics that of an average human male. This causes discrepancies between male and female patients, as physicians who use the model will not have the training essential for understanding the differences in female physiology. For example, blood vessels can create different output and pulse rates in response to emotional stress, as blood vessels in males will vasoconstrict while volumetric flow of blood increases in women [22]. General healthcare research tends to focus on the male physiology, which can often lead to improper treatment for female patients. As found in research, funding for health issues such as coronary artery disease is greater for men than women, although women are the population at higher risk [23]. Therefore, when marketing this project our group needs to ensure that consumers and users know the model primarily mimics the physiology of an average, healthy male. The team acknowledges that although our product is designed to mimic the human arterial system, it may not be a holistic representation of the various physiological functions between different individuals. Regardless, our system uses water and would not be able to take vessel blockages into account. In addition to gender and race, it is important to consider that adults over the age of 65 are more likely to suffer from cardiovascular diseases than younger people [24]. As stated before, our arterial model is designed to replicate the model of a healthy male, and thus may not replicate the physiology of older patients. In the future, it is essential that our system is representative of all genders, races, and any other factors that may contribute to differences in human anatomy or physiology.

7.4 Economic

The team does not foresee the project presenting any economic problems. On the contrary, the team successfully built a model that is able to replicate pulsatory flow through an arterial system model within a strict budget. This is an accomplishment when compared to the models that were found during review of existing literature. Our entire system was purchased for less money than the price of a new multi-channel peristaltic pump [13]. Designing an affordable system that effectively and accurately replicates pulsatile blood flow waveforms, gives people of a greater variety of economic statuses access to the tool. Surgeons at institutions with minimal resources will be able to practice intervention without the implications of a true surgical procedure, tissue engineers will be able to test the efficacy of cardiovascular stents or grafts with accurate hemodynamic stresses, and more students will be able to get a better understanding of how the cardiovascular system works beyond a textbook. Though a paywall still remains for this experience, the team's model makes it more accessible than it currently is. There is an initial possibility for our model to create a disparity within the healthcare system in regard to research on new intervention techniques for cardiovascular disease. As the model helps to replicate physiological conditions, researchers will be able to better design intervention systems, which will initially be expensive for patients in the healthcare system as are all new technologies. This will create a disparity as only a portion of people will be able to afford the new state of the art care. However, the priority first and foremost should be to improve the standard of cardiovascular intervention because it is currently in a place where revision is often required. Then, it should be made more accessible and affordable for the general public as quickly as possible. Based on this, the economic benefits of the model far outweigh the cost to purchase or assemble it.

7.5 Environmental

The team does not foresee any significant environmental impacts as a result of this project. However, we acknowledge that some of the components used in the project are mass produced in a factory setting, which produces a bulk of the world’s carbon emissions [25], [26]. The components produced in factories are inexpensive and widely available by vendors all over the world, which is a major benefit of this project. When possible, the team considered using sustainable materials. The artery model used in this project is made of polyethylene terephthalate (PET) which is recyclable. Overall, the potential to save lives with the development of improved interventional techniques far outweighs the minimal environmental impact from the project.

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Appendix A. Integrated System Arduino Code

```
double sensed_output, control_signal;
double setpoint;
double Kp = 0.7; //proportional gain
double Ki = 0; //integral gain
double Kd = 2; //derivative gain
int T = 10; //sample time in milliseconds (ms)
unsigned long last_time;
double total_error, last_error;
int max_control = 20;
int min_control = -20;
int pulseDelay = 100; //delay in ms

// Defining Analog Pin + Pressure
#define analogInPin  A0

float pressure = 0;
double t = 0;
double t_o = 0;
double t_1 = 0;
float set = 0;
int samples = 8;

#define pul 8
#define dir 10
#define enable 11

void setup()
{
  pinMode(pul, OUTPUT);
  pinMode(dir, OUTPUT);
  pinMode(enable, OUTPUT);

  // Begin Serial Monitor
  Serial.begin(2000000);

  digitalWrite(enable, LOW);
  delayMicroseconds(10);
  digitalWrite(dir, LOW);
  delayMicroseconds(10);
  digitalWrite(pul, LOW);
  delay(200);
}
```

```
void loop(){

  pressure = getPressure()*51.7;
  sensed_output = pressure;
  Serial.print("Pressure-mmHg:");
  Serial.println(pressure);

  t = millis()/1000.0;
  Serial.print("Time:");
  Serial.println(t);

 setpoint = 100 + 20*sin(2.0*3.14*t*8.0); //Sinusoidal function

 /*if(t-t_o < 0.5){ //Unit Step function
    setpoint = 80;
  }
  else if(t-t_o >= 0.5 && t-t_o <= 1.5){
    setpoint = 120;
  }
  else if(t-t_o > 0.75 && t-t_o < 2.0){
    setpoint = 80;
  }
  else {
    t_o = t;
  }*/

 /* if(t-t_o < 1){ //Waveform Function
    t_1 = t-t_o;
    setpoint = -1.5833e+4*pow(t_1,6) + 5.1533e+4*pow(t_1,5) - 6.4506e+4*pow(t_1,4)
+ 3.8601e+4*pow(t_1,3) - 1.1015e+4*pow(t_1,2)
    + 1.2294e+3*t_1 +68.0715;
  }
  else {
    t_o = t;
  } */

  Serial.print("Set-Pressure:");
  Serial.println(setpoint);

  //calls the PID function every T interval and outputs a control signal
  PID_Control();

  Serial.print("Control:");
  Serial.println(control_signal);
```

```
    if(control_signal < -10){
    pressureLow();
  }

  if(control_signal > 10){
    pressureHigh();
  }
}

void PID_Control(){

  unsigned long current_time = millis(); //returns the number of milliseconds
passed since the Arduino started running the program

  int delta_time = current_time - last_time; //delta time interval

  if (delta_time >= T){

  double error = setpoint - sensed_output;

  total_error += error; //accumulates the error - integral term
  if (total_error >= max_control) total_error = max_control;
  else if (total_error <= min_control) total_error = min_control;

  double delta_error = error - last_error; //difference of error for derivative
term

  control_signal = Kp*error + (Ki*T)*total_error + (Kd/T)*delta_error; //PID
control compute
  if (control_signal >= max_control) control_signal = max_control;
  else if (control_signal <= min_control) control_signal = min_control;

  last_error = error;
  last_time = current_time;
  }
}


float getPressure(void)
{
  float averagePSI = 0;
  int rawADC;
  float voltage;
  float psi;
```

```
  for(int i = 0; i < samples; i++)
  {
    rawADC = analogRead(analogInPin);
    voltage = (float) rawADC / 1024.0 * 5.0;
    psi = (voltage - 0.5) / 4.0 * 5.0;

    if(psi < 0.0 || psi > 5.0){
      continue;
      i--;
    }
    else{
    averagePSI += psi;
    delayMicroseconds(10);
    }
  }

  averagePSI = averagePSI/samples;
  return averagePSI;
}

void pressureLow(){
  digitalWrite(dir, LOW);
  delayMicroseconds(10);
  for(int i = 0; i<150; i++){
    digitalWrite(pul, HIGH);
    delayMicroseconds(pulseDelay);
    digitalWrite(pul, LOW);
    delayMicroseconds(pulseDelay);
  }
}

void pressureHigh(){
  digitalWrite(dir, HIGH);
  delayMicroseconds(10);
  for(int i = 0; i<150; i++){
    digitalWrite(pul, HIGH);
    delayMicroseconds(pulseDelay);
    digitalWrite(pul, LOW);
    delayMicroseconds(pulseDelay);
  }
}
```

Appendix B. MATLAB Code

```
clear;
X = readtable('C:\Users\rosen\OneDrive\Documents\MQP\unit_24.csv');
T_0 = 1; %Period in seconds

N = X(:,1).Variables;
X = X(:,2).Variables;
n = (length(X)/4)+1;
outPres = zeros(n,1);
setPres = zeros(n,1);
ctrl = zeros(n,1);
time = zeros(n,1);

o = 1;
s = 1;
c = 1;
t = 1;

for int = 1:length(X)
    if(contains(N(int),'mmHg'))
       outPres(o) = X(int);
       o = o+1;
    elseif(contains(N(int),'Set'))
       setPres(s) = X(int);
       s = s+1;
    elseif(contains(N(int),'Control'))
       ctrl(c) = X(int);
       c = c+1;
    elseif(contains(N(int),'Time'))
       time(t) = X(int);
       t = t+1;
    end
end

outPres(n) = [];
setPres(n) = [];
ctrl(n) = [];
time(n) = [];

ctrl = (ctrl/10.0);
time = time-time(1);

figure();
yyaxis left;
plot(time,outPres,'k--')
```

```
hold on
plot(time,setPres,'k-');
ylim([60 180]);
ylabel('Pressure(mmHg)');
yyaxis right;
plot(time,ctrl,'r-')
ylim([-10 10]);
legend('measured pressure','set pressure','control');
ylabel('Control Output');
xlabel('Time(s)');

figure();
plot(time,outPres,'k--')
hold on
plot(time,setPres,'k-');
ylim([60 140]);
xlim([12 13]);
xticks([12 12.16 12.33 12.5 12.67 12.83 13]);
xticklabels({'0','\pi/3','2\pi/3','\pi','4\pi/3','5\pi/3','2\pi'})
ylabel('Pressure(mmHg)');
legend('measured pressure','set pressure');
xlabel('Phase Angle (Radians)');

T = 0:0.01:1;
outNorm = zeros(length(T),1);
setNorm = zeros(length(T),1);
index = zeros(length(T),1);

outMax = zeros(length(T),1);
outMin = ones(length(T),1)*500;

l = length(outNorm);

outAvg = 0;
setAvg = 0;

for jnt = 1:length(outPres)
    if(time(jnt) > 1)
        time = time-1;
    end
    t = find(T == time(jnt));
    outNorm(t) = outNorm(t) +  outPres(jnt);
    setNorm(t) = setNorm(t) +  setPres(jnt);
    index(t) = index(t)+1;
```

```
    if(outPres(jnt) > outMax(t))
        outMax(t) = outPres(jnt);
    end
    if(outPres(jnt) < outMin(t))
        outMin(t) = outPres(jnt);
    end

    outAvg = outAvg + outPres(jnt);
    setAvg = setAvg + setPres(jnt);
end

outAvg = outAvg/jnt;
setAvg = setAvg/jnt;

x = T.';
y = outNorm./index;
y2 = setNorm./index;
knt = 1;

for int = 1:l
    if(isnan(y(knt)))
        x(knt) = [];
        y(knt) = [];
        y2(knt) = [];
        outMax(knt) = [];
        outMin(knt) = [];
    else
        knt = knt+1;
    end
end

fit = 12;
p1 = polyfit(x,y,fit);
p2 = polyfit(x,y2,fit);
p3 = polyfit(x,outMax,fit);
p4 = polyfit(x,outMin,fit);

x1 = linspace(0,1);
f1 = polyval(p1,x1);
f2 = polyval(p2,x1);
f3 = polyval(p3,x1);
f4 = polyval(p4,x1);

figure
hold on
plot(x1,f1,'k--')
plot(x1,f2,'k-')
```

```
plot(x1,f3,'k:')
plot(x1,f4,'k:')
ylabel('Pressure(mmHg)');
xlabel('Normalized Period (1Hz)');
legend('measured pressure','set pressure','measured extrema');
xticks([0 0.16 0.33 0.5 0.67 0.83 1]);
xticklabels({'0','\pi/3','2\pi/3','\pi','4\pi/3','5\pi/3','2\pi'})

e = 0;
for int = 1:l
    e = e + (outNorm(int) - setNorm(int))^2;
end

disp(['RSME: ',num2str(sqrt(e/l)/length(outMax)),' mmHg']);
disp(['NRSME: ',num2str(sqrt(e/l)/setAvg),'%']);
disp(['Mean Offset: ',num2str(outAvg - setAvg),' mmHg']);
disp(['Magnitude Response: ',num2str(outAvg/setAvg)]);

sum = 0;
for int = 1:length(outMax)
    sum = sum+outMax(int);
end
sum = sum/length(outMax);
disp(['Max Response: ',num2str(sum/setAvg)]);

phaseDelay =(find(f2 == max(f2))) - (find(f1 == max(f1)));

sum = 0;
for int = -phaseDelay+1:length(x1)
    sum = sum + (f2(int)/f1(int-3));
end
sum = sum/(int+phaseDelay-1);

disp(['Phase Adjusted Mean Offset: ',num2str(sum),' mmHg']);

phaseDelay =(x1(find(f2 == max(f2)))) - (x1(find(f1 == max(f1))));

disp(['Phase Delay: ',num2str(phaseDelay),' s']);
```

Appendix C. Bill of Materials

Vendor	Item Description:	Specifications	Qty:	Unit Price:	Price:
Mc-Master Carr	Firm Polyurethane Tubing for Air and Water	Clear, 10ft	2	$13.10	$26.20
Mc-Master Carr	Plastic Barbed Tube Fitting for Air and Water		1	$10.57	$10.57
Mc-Master Carr	Plastic Storage Container		1	$7.07	$7.07
Mc-Master Carr	Thick-Wall Clear Blue Unthreaded PVC Pipe for Water	8ft, 1/4" NPT	1	$15.00	$15.00
Mc-Master Carr	Plastic Barbed Tube Fitting for Air and Water		1	$5.84	$5.84
Mc-Master Carr	Plastic Barbed Tube Fitting for Air and Water		1	$5.67	$5.67
Mc-Master Carr	Clear PVC Pipe Fitting for Water		1	$17.76	$17.76
Mc-Master Carr	Thick-Wall PVC Plastic Pipe Fitting for Water		2	$10.90	$21.80
Mc-Master Carr	Development Board		2	$29.57	$59.14
Mc-Master Carr	Aluminum Washer		1	$9.62	$9.62
Mc-Master Carr	Medium-Strength Steel Hex Nut		1	$13.73	$13.73
Mc-Master Carr	Back-Pressure-Regulating Valve for Water and Oil	15-20psi	1	$291.51	$291.51
Mc-Master Carr	Round Body Air Cylinder		1	$144.74	$144.74
Amazon	Linear Stage Actuator		1	$87.80	$87.80
Amazon	STEPPERONLINE CNC Stepper Motor Driver		1	$28.99	$28.99
Home Depot	Brass Reducing Fittings		1	$9.29	$9.29
Home Depot	PVC Sheet		1	$39.69	$39.69
Home Depot	Socket Caps and Nuts		1	$10.87	$10.87
Home Depot	1/4" NPT Ball Valve		1	$20.25	$20.25
Amazon	J-B Weld		1	$6.25	$6.25
Amazon	0-10 psi Pressure Transducer		2	$18.37	$36.74
Amazon	12 Volt 15 Amp Power Supply		1	$32.29	$32.29
Amazon	12 Volt 6 Amp Power Supply		1	$12.70	$12.70
Amazon	Motor Driver Controller Board Modules		1	$22.82	$22.82

Mc-Master Carr	Stepper Motor, NEMA 23, 352.5 in.-oz. Maximum Holding Torque		1	$184.64	$184.64
Amazon	KOLERFLO 12 Volt Water Pump	5.5 GPM 70 PSI	1	$89.99	$89.99
Amazon	1/4" Check Valve One Way Inline		1	$9.99	$9.99
N/A	Estimated cost for 3D Printed Parts		1	$100.00	$100.00
Final Design Total					**$1,320.96**
Items purchased but not used in final design					$581.89
Total					**$1,902.85**