University of Wisconsin - Madison
Department of Biomedical Engineering

Arterial Actuator

Final Report

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Abstract
In an effort to diagnose and prevent cardiovascular disease, the ultimate goal of this project is to design a system that can non-invasively measure blood pressure and arterial stiffness on a single artery. Through preliminary pressure sensor research, it was determined that the artery used for data collection does not significantly affect output. Therefore, the radial artery was chosen due to its ease of access. The final design incorporates a push/pull solenoid as the actuator and a piezoelectric sensor mounted inside of an inflatable cuff for data collection. A rigid arm holder further stabilizes the system. The signal from the sensor was amplified, filtered, and exported to a Java program for data acquisition and display. Quantitative blood pressures, pulse pressure, and pulse rates were obtained. Systolic, diastolic, and pulse pressure deviated from the auscultatory method by 10.6374, 8.6580, and 14.7066 mm Hg on average, respectively. In addition, qualitative arterial stiffness data and average response lengths were analyzed after actuation.

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I. Problem Statement

Cardiovascular disease is one of the top killers in today's society. Blood pressure and arterial stiffness are indicators of cardiovascular health. Currently, blood pressure is measured via sphygmomanometry and arterial stiffness via arterial tonometry. Although effective, the speed and accuracy of these methods can be improved. The goal is to design a system, comprised of a piezoelectric pressure sensor, an actuator, and a stabilizing structure, that can quantitatively measure blood pressure and arterial stiffness on a single artery.

II. Background

Arterial blood pressure is the key driving force for propelling blood to the body's tissues (Sherwood et al., 2005). Blood pressure is highly regulated in order to ensure sufficient nutrient and oxygen supply to the body. The maximum pressure exerted in the arteries, known as systolic pressure, occurs as the heart contracts. In contrast, the minimum pressure, known as the diastolic pressure, occurs with the relaxation of the heart between contractions (Sherwood et al, 2005). Blood pressure is usually reported as systolic pressure over diastolic pressure, which is approximately 120/80 mm Hg in a healthy adult. Furthermore, pulse pressure is another key measurement used to determine heart health. Pulse pressure is the difference between the systolic and diastolic pressures, which can be used to determine the cardiac output of the heart (O'Rourke et al., 1999). High pulse pressure is an indication of and may lead to artery damage. Normally, resting pulse pressure for a healthy adult is 40 mm Hg.

Arterial stiffness is the elasticity of the arterial walls. The arteries stiffen as a result of age, atherosclerosis, and fraying of the elastic fibers in the arterial walls. Atherosclerosis is caused by plaque buildup, which can lead to heart attack, stroke, and other cardiovascular diseases (O'Rourke et al., 2002). As arteries stiffen, more force is required for them to expand properly, forcing the heart to work harder to pump blood to the body.

Current methods

Main methods of measuring arterial blood pressure include the auscultatory and oscillometric methods. The auscultatory method uses a mercury sphygmomanometer (i.e. blood pressure cuff) to measure blood pressure by the Korotkoff sound technique. The auscultatory method is performed by placing the stethoscope below the blood pressure cuff, and then inflating the sphygmomanometer to about 180 mm Hg. After inflation, air pressure within the cuff is slowly released. **Figure 1** shows the setup of measuring arterial blood pressure using the sphygmomanometer and stethoscope. The pressure displayed on the sphygmomanometer when the first heartbeat is heard is the systolic blood pressure. The diastolic blood pressure occurs when heartbeats are no longer audible. These sounds are caused by the turbulent flow that exists due to occlusion in the arteries when constricted to a level between systolic and diastolic pressures (Widmaier et al., 2008).

Figure 1: Measuring arterial blood pressure via the auscultatory method

Image courtesy of Acupuncture and Herb:
http://www.acupuncturebrooklyn.com/uncategorized/hypertension-epidemic-causedd-by-wrong-bp-cuff- size-karen-vaughan

On the other hand, the oscillometric method uses an array of electronic transducers to detect the degree of oscillations caused by the blood. In this method, the pressures are recorded based on the oscillations that occur during turbulent flow rather than the sounds. While the oscillometric method is easier and faster, some studies indicate that it is less precise when compared to the auscultatory method (Park et al., 2001). **Figure 2** shows a blood pressure monitor that uses the oscillometric method.

Figure 2: Blood pressure monitor that uses the oscillometric method

Image courtesy of Blood Pressure Monitors:
http://blood-pressure-monitors.thebighealth.com

The most effective way currently used to measure arterial stiffness is via pulse wave velocity (PWV). PWV is calculated based on the transition time and distance traveled by the pulse between the carotid artery and femoral artery (O'Rourke et al., 2002). Ways to measure PWV include pressure-sensitive transducers, Doppler ultrasound, and applanation tonometry. After collecting data via PWV, arterial stiffness is calculated via the Moens-Korteweg equation: $\mathrm{PWV} = \sqrt{(Eh/2\rho R)}$. In this equation, E is Young's modulus (elasticity/stiffness) of the arterial wall, h is wall thickness, R is arterial radius at the end of diastole, and ρ is blood density (Oliver and Webb, 2003).

Competing products for our design that currently exist in the market include the Pulse Pen, which is composed of a tonometer and an integrated electrocardiogram unit. A single operator can run the pen, which measures central aortic pressure and records pressure waveforms using applanation tonometry (www.pulsepen.com). Also, Jeffrey C. Petzke and Dennis E. Bahr patented a similar device to that of this project on December 16, 1975, titled "Blood Pressure Measuring Apparatus" (Patent 3,926,179).

III. Preliminary Designs

Actuator for measuring arterial stiffness

One of the primary goals of the project is to design and fabricate an actuator that is able to produce a step input on the radial artery. This will allow for the assessment of arterial stiffness. The four proposed mechanisms for the actuator were spring-loaded, air-loaded, air jet, and protrusion-in-a-cuff designs.

The spring-loaded design, similar to the release of a pen or a solder remover, uses a spring force to actuate a piston onto the artery. This design incorporates variable spring compressions to yield a number of lengths of release. This provides adjustability and accounts for differences in patient body composition and artery location. The air-loaded design (**Figure 3**) uses a similar actuating mechanism as the spring-loaded design: however, it uses compressed air to drive the

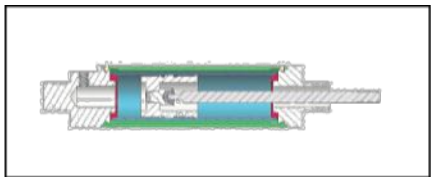

Figure 3: An air-loaded actuator

Image courtesy of Direct Industry: http://www.directindustry.com

piston and provide the necessary compression. This actuator also allows for individual adjustments.

The air jet design incorporates a needle, attached to a source of compressed air, which can concentrate air in a small area. The force of the air stream itself is then used to put the necessary pressure on the artery without the use of an actuating object. Lastly, the protrusion-in-a-cuff design is comprised of an object permanently secured to the interior of an inflatable cuff. The object will move with the cuff and thereby compresses the artery with cuff inflation.

Overall system stability and assembly

In order to assemble the components of the system, a stabilizing system is needed to steady the wrist and eliminate motion artifacts from measurements on the radial artery. The stabilizing system needs to incorporate the actuator and pressure sensor, which must be positioned directly above the artery. Three preliminary designs were proposed for the stabilizing system: a portable brace, a mounted sleeve, and a surface strap.

In the portable brace design, an adjustable Velcro strap holds the stabilizing system together. This design allows the user to move their arm freely but keeps their wrist stable, similar to a wrist brace used after injury. The actuator and pressure sensor are locked into place once their parameters are set. The next design, the mounted sleeve, involves an adjustable cuff that is secured to a surface (i.e. table or counter) that disallows patient movement. The wrist is inserted into the cuff, which is then adjusted to fit properly. The actuator and sensor components can either be a part of the cuff or secured independently. Finally, the surface strap design is similar to the mounted sleeve in that all movement is prohibited. Straps (such as Velcro) are attached to a surface and tightened around the wrist and forearm, while the components of the system are attached to the surface next to the wrist.

IV. Preliminary Design Evaluation

Each preliminary design had its own strengths and weaknesses, which were evaluated with a design matrix. One design matrix was compiled to evaluate the different actuating mechanisms, while another considered the overall system stability and assembly methods.

Actuator for measuring arterial stiffness

The four actuating mechanisms were rated on a variety of design criteria. These aspects were selected because of their importance in an effective design. It was determined that certain criteria, such as performance, adjustability, and patient comfort, were more significant and therefore were given greater emphasis. The scores for each design in each category were summed to give a total score out of 100, shown in **Table 1**.

Table 1: A design matrix displaying the design criteria and weight values of the four actuating designs

Weight	Criteria	Spring-Loaded	Air-Loaded	Air Jet	Protrusion in Cuff
30	Performance	24	28	14	14
20	Adjustability	12	17	17	15
20	Patient Comfort	13	13	17	14
12	Ease of Fabrication	9	8	11	11
10	Durability	7	6	9	8
8	Size	6	5	5	7
100	**Total**	**71**	**77**	**73**	**69**

Performance is defined as the ability of the actuator to give an exact step input to the artery. It was given a weight of 30 points in the matrix, designating it as the most important category, because it determines the ultimate effectiveness of the actuating mechanism. The spring-loaded and air-loaded designs scored highest, as it was speculated that the air jet and protrusion ideas would not work in an instantaneous on-off fashion.

Adjustability is another concern for the actuator. As a result of differing patient body compositions, artery locations, and artery depths, the design must be customizable to perform on a wide variety of patients. The two air-based designs, air-loaded and air jet, were determined to be most effective in this area because air pressure can be varied more easily and potentially more precisely than the other approaches. Additionally, patient comfort was weighted heavily in order for feasibility of clinical implementation.

After creating and evaluating the design matrix, the air-loaded design was determined to be the most effective option for the actuator. One flaw of this design matrix, however, was its subjectivity. At the time of evaluation, the ratings given were based on team speculation of how the design would fare in that area. Therefore, experimental testing and further research was necessary to ensure that the most effective design was selected.

Overall system stability and assembly

Much like the actuating mechanism, the portable brace, mounted sleeve, and surface strap designs were evaluated using a design matrix (**Table 2**) to determine the best selection for an overall system stability and assembly method.

Table 2: A design matrix evaluation of the three system stability and assembly methods

Weight	Criteria	Portable Brace	Mounted Sleeve	Surface Strap
30	Patient Comfort	24	21	15
20	Stability	10	18	16
20	Ease of Fabrication	17	16	19
20	Ease of Clinical Use	18	16	16
10	Aesthetics	6	7	5
100	**Total**	**75**	**78**	**71**

Patient comfort stood paramount in selecting the overall system design. The current methods of measurement are relatively comfortable processes. Therefore, the new design must be equally as comfortable, if not more. The design cannot induce stress

because it causes variations in the reading of blood pressure and pulse rate. It was determined that the portable brace is the most comfortable method for the patient, followed by the mounted sleeve.

Stability was another substantial concern in this design selection; this system must be able to hold the patient's arm relatively still. If the sensor is moved during testing, motion artifacts will disrupt the output. The portable brace fell short in this category despite standing out in patient comfort. In addition to comfort and stability, the ease of clinical use and fabrication were also considered significant and weighed heavily. After careful consideration of each design's strengths and weaknesses, the mounted sleeve design scored the highest in the design matrix.

V. Final Design

The final system is comprised of several components: a pressure sensor, amplifying circuit, actuator, and stabilizing structure. The following final components were chosen after further research and reconsideration of preliminary designs.

The first component, a piezoelectric sensor, was provided by the biomedical engineering department and implemented in the final design. A piezoelectric sensor is comprised of a crystalline material that accumulates charge and generates a potential when deformed (Piezocryst, 2005). The pressure sensor outputs a voltage waveform, which can be used to display a patient's cardiac activity on an oscilloscope. The output of the sensor is amplified and filtered with a LT1920 op-amp and referenced to 2.5 volts. This amplifies the output by a factor of 10. An original circuit was constructed on a breadboard; however, its filters eliminated 0.5 - 40 Hz signals from the output. This led to a new circuit design, which is wire-wrapped to a microcontroller board with USB output (**Figure 4**). The net list for this

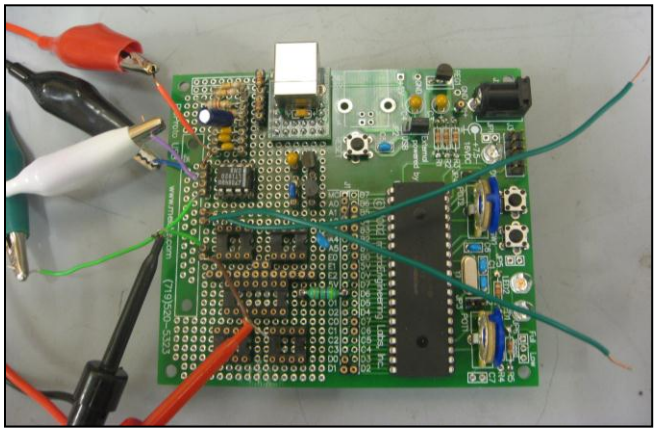

Figure 4: The amplifying and filtering circuit wire-wrapped to a microcontroller board

wire-wrapping can be seen in **Appendix D**. A detailed schematic of this final circuit can be seen in **Figure 5**. The USB port allows for the output to be displayed on a computer using Java data acquisition software developed by fellow UW-Madison biomedical engineering student Matthew Bollom.

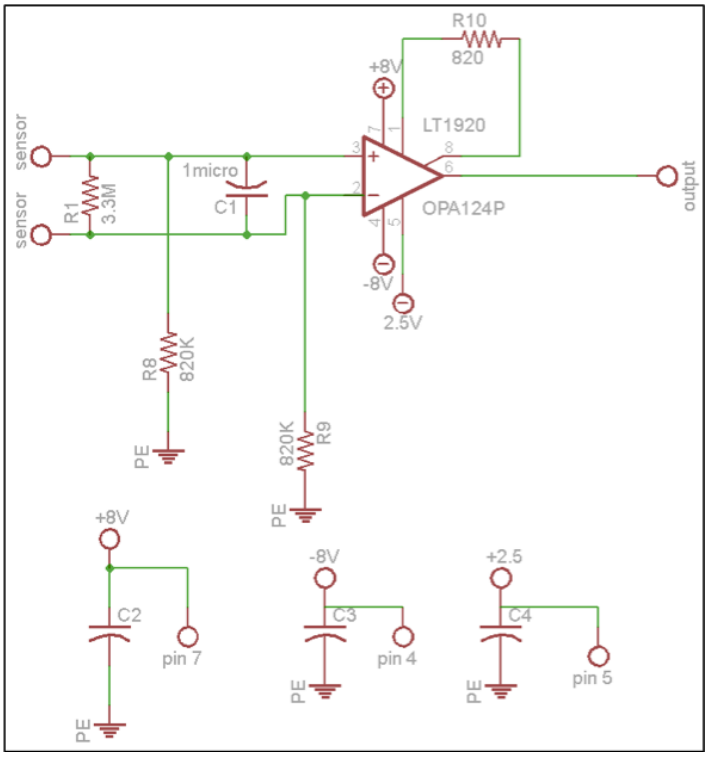

Figure 5: Schematic of the final amplifying and filtering circuit built on Eagle.

The sensor itself is placed within a 17.7927 mm × 24.6634 mm × 6.5024 mm block, custom fabricated from aluminum. The purpose of this block is to increase the surface area in contact with the patient's arm, allowing for effective compression of the artery while minimizing discomfort. The aluminum block and sensor unit is integrated into a modified sphygmomanometer cuff (**Figure 6**) in order to keep the sensor in place and compress the sensor into the artery to the point where it is occluded halfway. When this occurs, the pressure applied to the artery by the compression force is equal to the pressure applied by the blood inside the artery (i.e. blood pressure). In order to fit on the patient's wrist, the cuff was modified to be 6.429 cm in width.

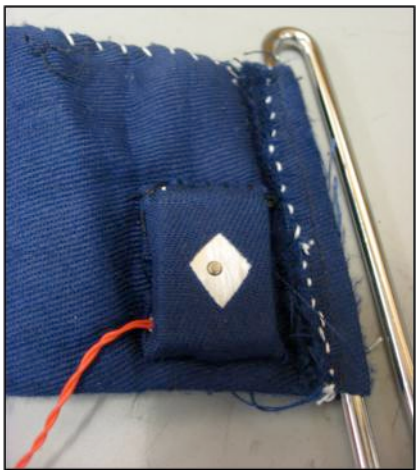

Figure 6: The sensor and aluminum block system attached to the inflatable cuff

The cuff used is adjustable for each patient during testing. It is a combination of the protrusion-in-a-cuff and portable brace ideas. Initially, the cuff was to be secured to the stabilizer, but it was decided that adjustability is greater when the sensor is not fixed in place. By keeping the cuff independent, it can be rotated and moved so that the sensor is accurately over the artery. The inflatable cuff is shown in **Figure 7**.

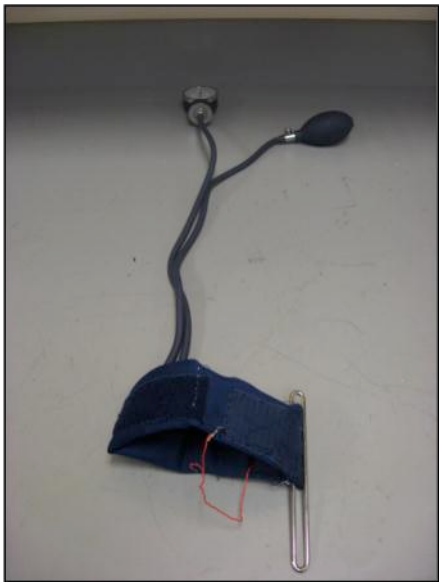

Figure 7: The modified blood pressure cuff used for the wrist

Another component of the system is the actuator, which is used to induce a step input force on the artery. Theoretically, this is an instantaneous hit. Once this force is applied, the response of the artery can be observed on the pressure waveform and the arterial stiffness can be determined. The longer it takes the artery to return to its normal state, the stiffer the artery.

A combination linear push/pull solenoid actuator was chosen for the actuator. Originally, the design was to incorporate a diaphragm air actuator that pushes a piston out when air is compressed within the cylinder, and retracts the piston via an internal spring when air is released. The diaphragm air actuator was not used because air compression is not a practical design for clinical use. Furthermore, the decision to use a solenoid actuator was made after consulting with biomedical engineering Ph. D. student Dennis Bahr, who worked on a similar project previously. The push-pull solenoid actuator, model 6873K12, was ordered from McMaster-Carr (**Figure 8**). It can receive up to 24V DC, has a rod diameter of 3.175 mm, and can provide up to 30 oz. of force at a 3.175 mm stroke (maximum stroke length is 7.874 mm). When a quick pulse of 24V DC is applied to the solenoid, the solenoid extends the rod out into the artery. An added compression spring with an outer diameter of 10.716 mm and a length of approximately 9.728 mm allows for quick retraction when the voltage is no longer applied, creating the step input desired for actuation. This actuator is secured to the stabilizing unit using a light fixture whip, a flexible tube that allows for the operator to adjust the location of the actuator so it can hit the artery accurately while simultaneously providing stability to the set-up.

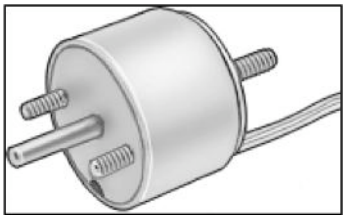

Figure 8: The push/pull solenoid used to actuate

Image courtesy of: mcmaster-carr.com.

Lastly, the stabilizing structure was constructed to immobilize the patient's arm during testing, as shown in **Figure 9**. The structure incorporates all design components for ease of use. The final stabilizing mechanism is a modified combination of the preliminary mounted sleeve and portable brace designs. The patient's arm is placed on the stabilizing unit during data collection. This unit is constructed from a four-inch diameter PVC pipe mounted on a wood base and support at a 15 degree angle from the horizontal. The end of this piece was connected by a 45 degree joint and the pipe was cut longitudinally to remove the semi-cylindrical top portion. Two Velcro straps are then incorporated to hold the arm and hand in place. This unit allows the patient's arm to remain immobilized, reducing motion artifacts and maximizing artery access. The final prototype costs for each component can be found in Table 1 of the appendix.

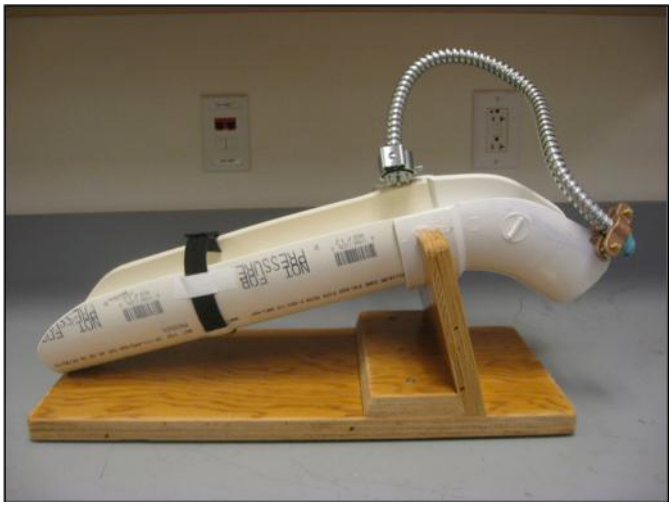

Figure 9: The stabilizing structure with the solenoid connected via light fixture whip

VI. Testing Procedure

Before results could be obtained, the sensor needed to be calibrated. Since there is no existing data sheet for the sensor, the exact conversion factor to convert voltage to pressure is not known. In order to calibrate the sensor and determine this value, masses of different sizes ranging from 2 to 10 grams were placed on the sensor and the voltage output was measured on an oscilloscope. Then, the pressures applied to the sensor by each mass were calculated and plotted versus the output relative to the reference voltage of 2.5. The slope of this linear relationship was used as the conversion factor for the sensor to convert voltage to pressure data.

Once the sensor was calibrated, it was reattached to the cuff and the corresponding circuit. To test the system, the cuff was placed on one of the group member's wrists and the waveform produced on the oscilloscope was recorded. The complete system setup is shown in **Figure 10** and a close-up of the use of the system is shown in **Figure 11**. A normal pulse waveform was recorded as the control when the systolic peaks were at maximum amplitude. At the point of maximum amplitude, the pressure reading on the gauge was recorded so the tests could be repeated more easily for each individual. The data from the control graph was exported to Microsoft Excel where the pulse rate and blood pressure (systolic, diastolic, and pulse) were calculated.

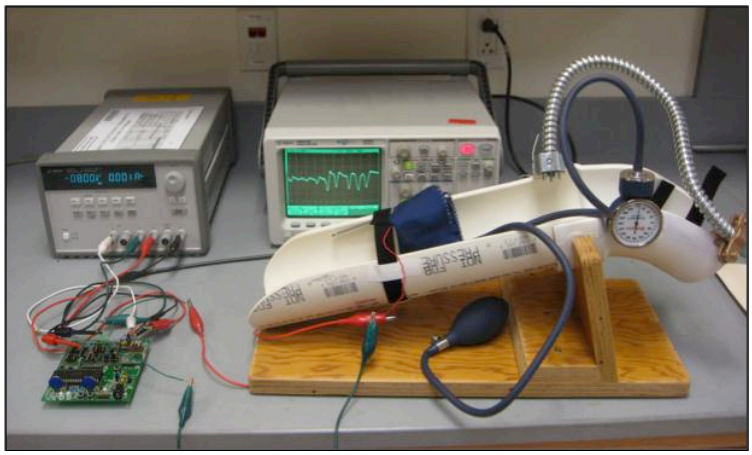

Figure 10: The complete testing setup – the stabilizing structure, the solenoid, the sensor and cuff, the amplifying circuit, the oscilloscope, and the power supply

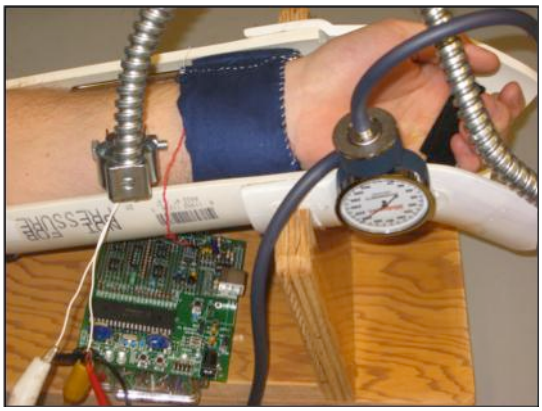

Figure 11: Position of the cuff on the wrist and the arm on the stabilizing structure w/the amplifying circuit placed on the wooden support.

Along with the control waveform, a waveform with actuation was recorded. On this waveform the input from the power supplied to the solenoid was also displayed. This was done in order to find the time delay between the impulse and artery response. Also, the time in which the 24V DC signal is supplied to the solenoid was calculated so that a Java program could potentially provide voltage to the solenoid for the same period of time when the system is fully automated. To give insight to the stiffness of the artery, the duration of the artery response was calculated. The duration is the time from the initial response of the artery to the time when the artery returns to its normal state. With stiffer arteries, the duration of the response will be shorter with smaller amplitude. If the arteries are more elastic, the response duration is longer with greater amplitude because the artery wall is more flexible. All of the tests explained above were performed

on each group member and on advisor Amit Nimunkar. **Figure 12** shows the testing procedure implemented on Evan Flink by fellow team member Clara Chow.

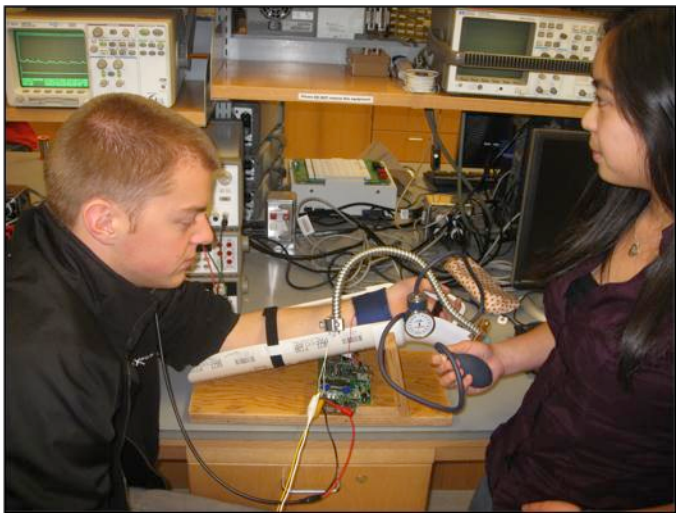

Figure 12: Team member Clara Chow determines the maximum signal from Evan Flink's arteries

The final step in the testing process is comparing the results found using the device with that of a standard blood pressure cuff. The most widely accepted way to measure blood pressure is using a sphygmomanometer, so if the results of this project are close to that found by a blood pressure cuff, the system is accurate and has the potential for clinical use.

VII. Results and Discussion

Theoretically, the calibration procedure for the sensor would give reasonable results that could be easily used to convert voltage to pressure data. However, significant flaws exist in this procedure. Since the sensor only measures dynamic pressure, the masses had to be dropped from a slight distance above the sensor. Not only did this create variability in the calibration, but impact forces are significantly greater than static ones. Additionally, the geometry of both the masses and the sensor pose additional problems. The masses used have concave bases, and the tip of the sensor has a raised ridge with a 1.58 mm outer diameter and 0.15 mm thickness. The pressure applied by a mass is inversely proportional to the surface area its force is spread over. It was assumed that this surface area was equal to that of this ridge, but there is no way to actually determine if this was true during calibration.

When the first few pressure waveforms were obtained and converted using the preliminary calibration constant, it was apparent that the form and proportions of the data were correct, but the pressure values were significantly lower than expected. It was determined that the force applied to the sensor should be scaled by a factor of five, in order to adjust for the problems that occurred in calibration. Doing this consistently corrected the final results to more salient values. The conversion factor used for all subsequent measurements is 1357.1 mm Hg/V. The calibration curve used to generate this constant can be seen in **Figure 13.**

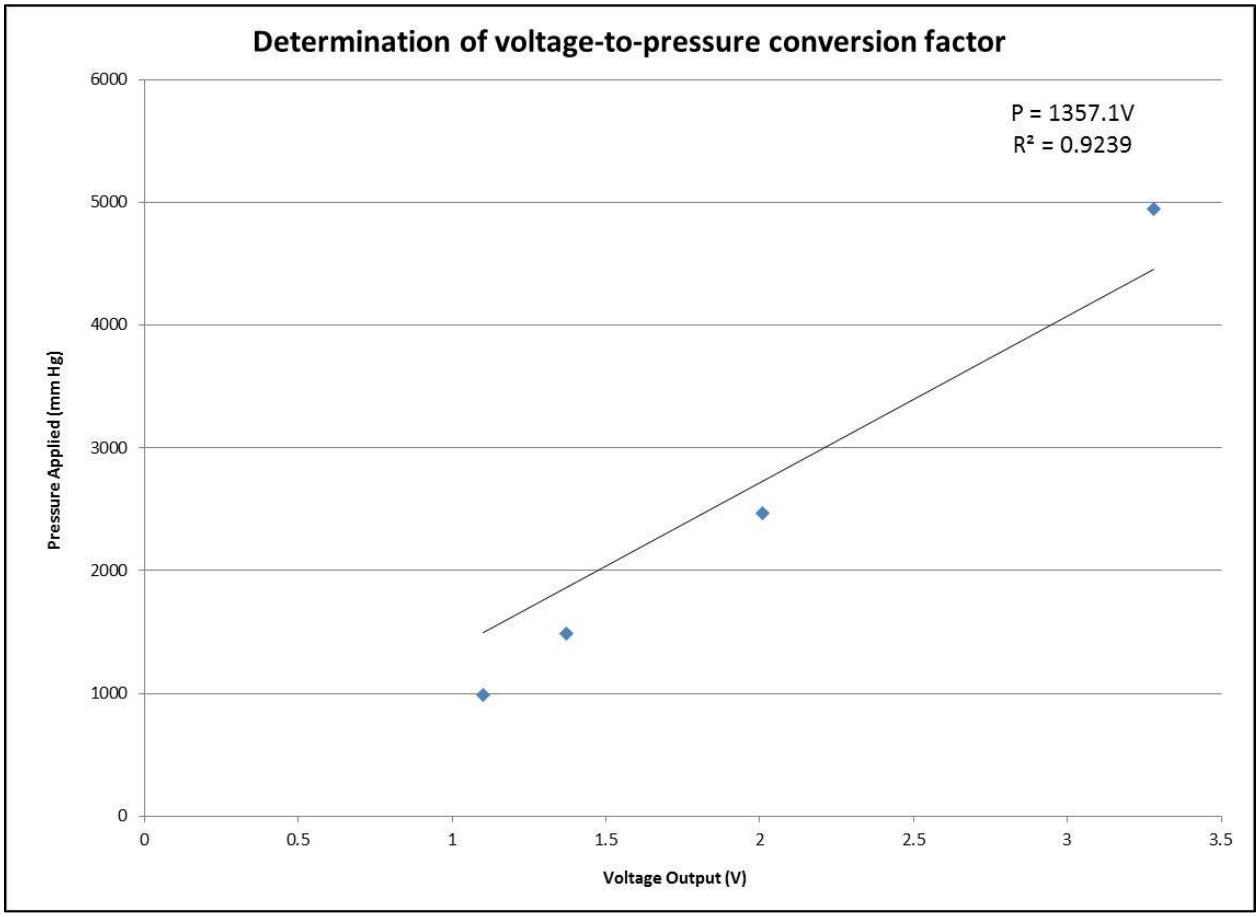

Figure 13: Pressure applied (corrected) vs. voltage output during sensor calibration – the slope of this line is the calibration constant used to convert between the two measurements

It is important to note that the voltage used to convert to pressure cannot be taken from a reference of exactly 2.5 V like during calibration. The reference value varies slightly over time and from subject-to-subject since there is a low frequency waveform component that is based on the subject's breathing pattern. Therefore, the reference was recorded as the baseline seen on the oscilloscope before each measurement. An example

of a blood pressure vs. time waveform recorded from testing can be seen below in **Figure 14**, and all plots for all five subjects can be seen in **Appendix E**.

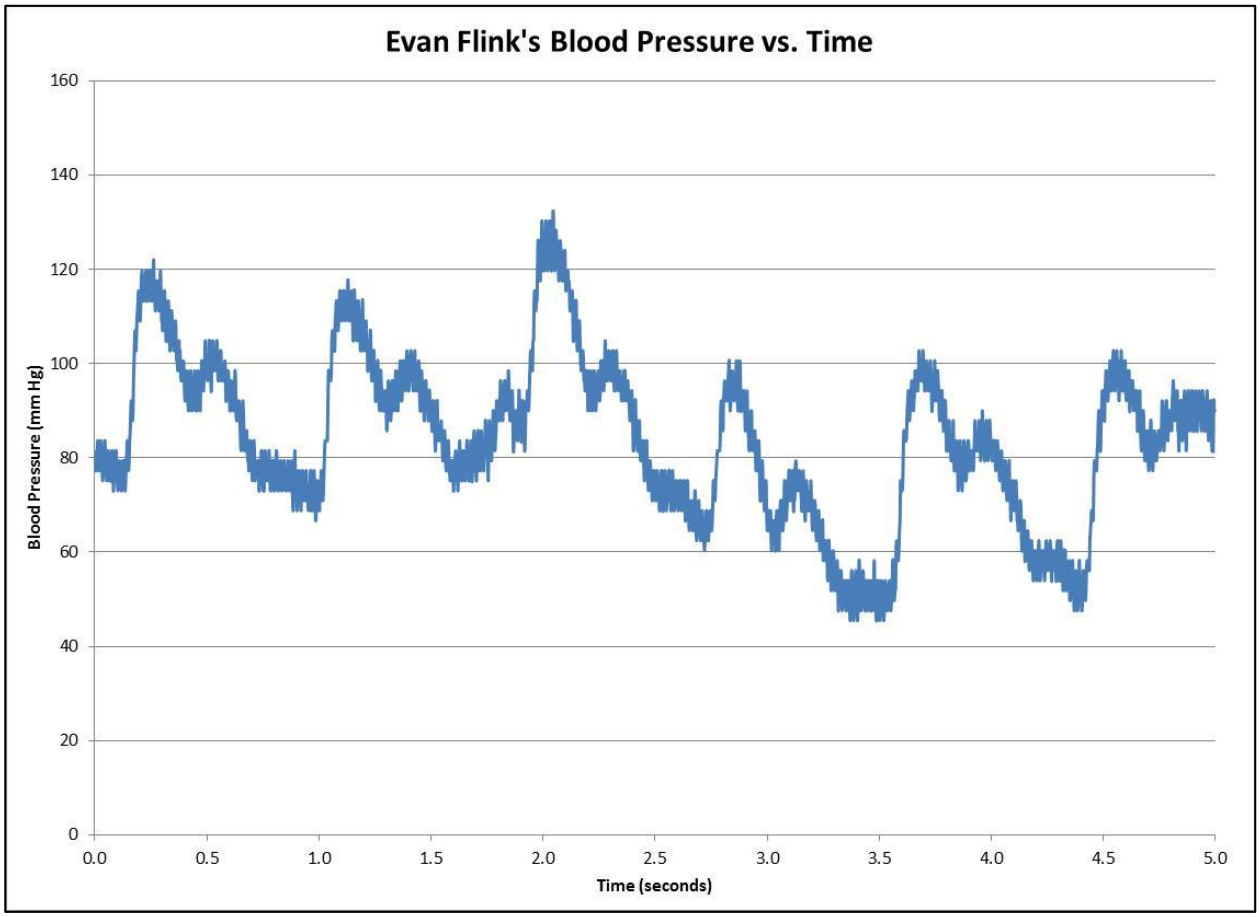

Figure 14: Team member Evan Flink's blood pressure waveform

In each plot, data was recorded for five seconds; five to seven heartbeats generally occurred during that period. When the average time between heartbeats for each subject is considered for each of the five participants, it corresponds to an average pulse rate of 49.4338 beats per minute. This is interesting physiologically because the average human has a pulse rate of 60-100 beats per minute (Widmaier et al., 2008). Perhaps the constriction of the radial artery during testing reduces the pulse rate through it.

The systolic and diastolic pressures can clearly be seen for each heartbeat, as well as the dicrotic notch where pressure briefly increases as the aortic semilunar valve closes in the heart (Widmaier et al., 2008). For each pulse wave, systolic and diastolic pressures were recorded and averaged for each subject. These values, as well as the

pulse pressure, were compared with the values recorded using the auscultatory method (**Figure 15**).

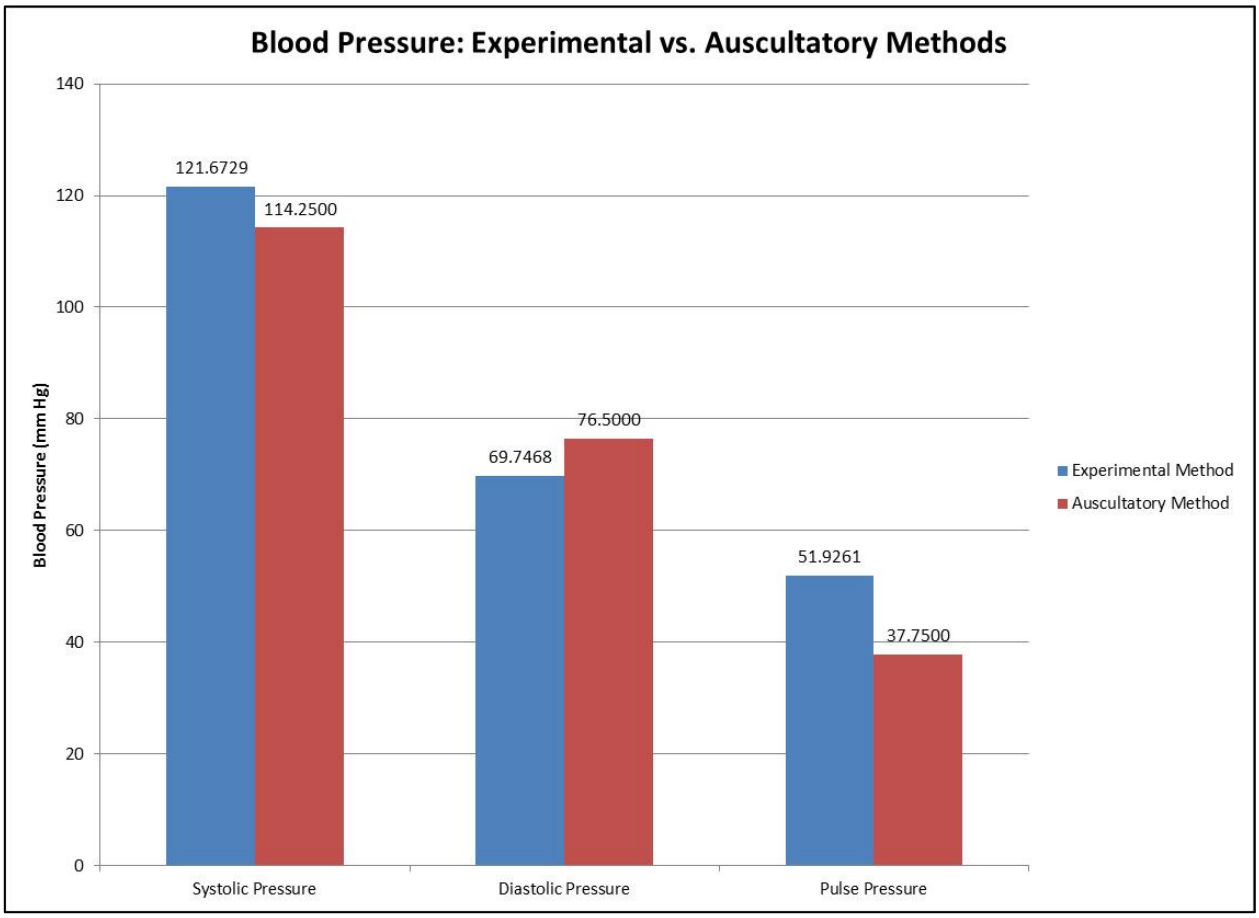

Figure 15: A comparison of average blood pressure values recorded using both the team's experimental method and the standard auscultatory method

The average deviations from the auscultatory method for systolic, diastolic, and pulse pressure were 10.6374, 8.6580, and 14.7066, respectively. The fact that the pulse pressure varies the most is likely due to the noise that occurs on each blood pressure vs. time plot, causing systolic pressure to appear too high and diastolic pressure to appear too low. Regardless, the accuracy of these experimental values is promising.

In addition to blood pressures and pulse rates, the device is also capable of actuating an artery to view its response. Blood pressure vs. time was plotted a second time for all five subjects, but in this trial the actuator struck the artery twice. It should be noted that the operator attempted to strike the artery directly after the dicrotic notch in the waveform; this was done to increase consistency and quality of the data. While the notch can be seen in most of the pressure waveforms, it is most obvious in an

electrocardiogram. Because of this, an ECG waveform will be displayed in the same Java data acquisition window as the blood pressure waveform, given that the subject is connected properly. **Figure 16** shows the resulting blood pressure waveform with

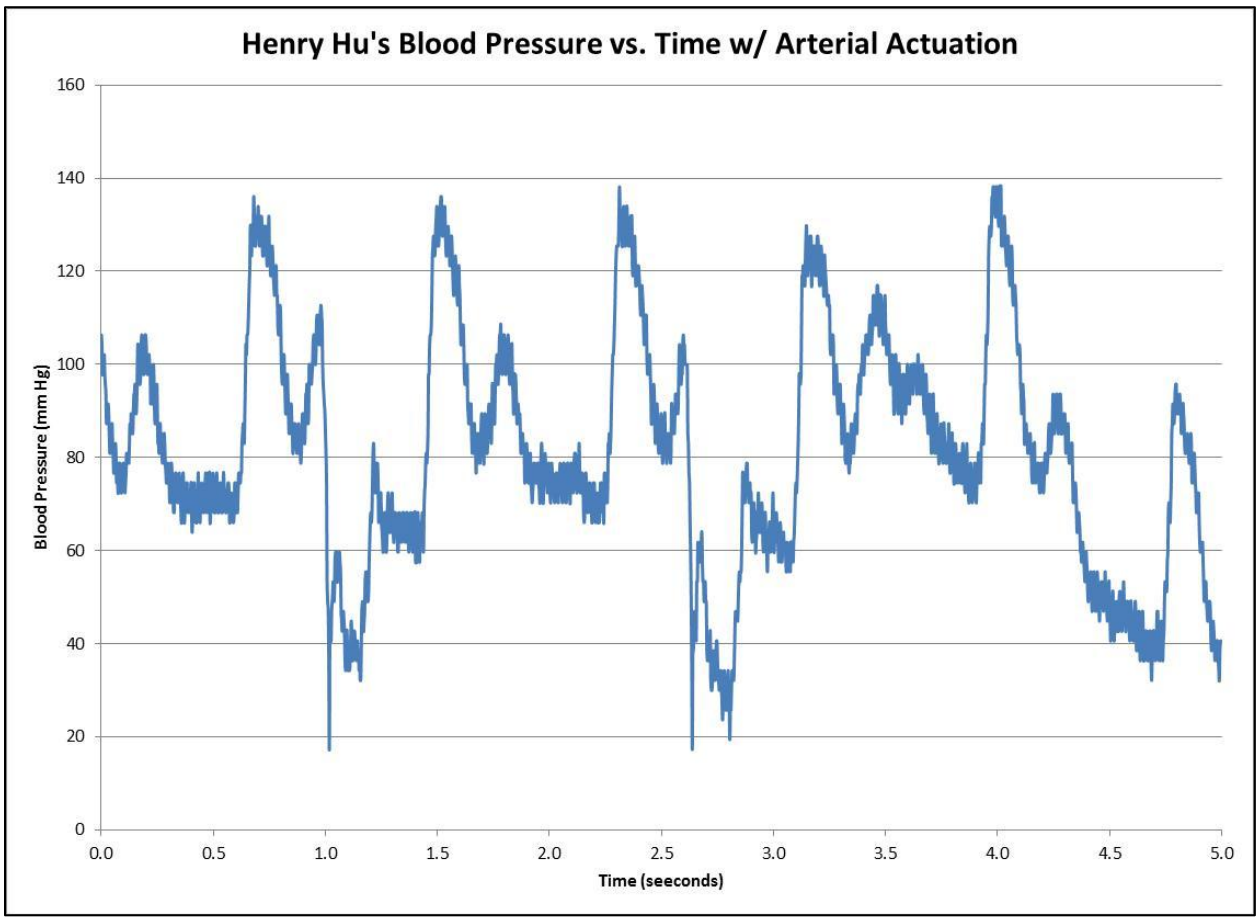

Figure 16: Team member Henry Hu's blood pressure waveform w/ arterial actuation

arterial actuation that resulted from one of the tests; the remainder can be found in **Appendix E**.

A great deal of information can be derived from waveforms such as these. The timing of the 24V signal to the actuator was recorded in addition to this waveform. It was determined that the average human-controlled pulse lasted 0.1616 seconds. This is useful because the actuation of the solenoid is to be automated in the future. The 24V timing was also compared with the timing of the arterial response, and averaged for all five subjects. There was only a 0.0041 second delay on average between 24V input and arterial response; this means that the actuator was close enough to the artery and has effective speed to have an almost instantaneous response.

In addition to visually comparing the deflection of the waveforms in response to actuation, the duration of arterial response and deflection of the waveform can be used to more quantitatively evaluate arterial stiffness. The average response duration for our subjects was 0.2047 seconds. This data was quite consistent except for one outlier of 0.0563 seconds, which is likely due to the fact that during that particular test the subject's artery may not have actually been hit directly by the actuator. All five subjects were fairly young and in generally good health, so it makes sense that the numbers related to arterial stiffness should be similar. If older individuals or subjects with poor cardiovascular health were tested, more meaningful results could potentially be obtained. In contrast, the deflection data was wildly inconsistent and could not be analyzed. This could be due to the fact that a 24V impulse was used at all times and distance from the skin was not necessarily taken into account when actuating the artery, so the levels of compression could have varied immensely. All of the key data derived from both waveforms that is discussed here can be seen in more detail in **Appendix F**.

VIII. Semester Summary

The proposed timeline from the beginning of the semester can be found in **Appendix B**. The order of the project development was followed but complications arose which hindered project progress. First, ultrasound was initially a component of the design; however, the available ultrasound machine was not functioning and could not be fixed. This resulted in a deviation from the proposed plan to compensate for the elimination of ultrasound. Moreover, there were issues with ordering materials because of the discovery of the patent, which consequently delayed prototyping and testing. Also, the amplifying circuit was revised multiple times that required repeated sensor calibration, data acquisition, and analysis.

The expenditures for this project can be found in **Appendix C**. The budget allocated at the beginning of the semester was somewhat limited, which influenced the choice of design components. The sensor was provided by the department and some recycled materials were used to fabricate the system in order to reduce cost. This resulted in a final prototype cost of $83.87, making the system quite affordable.

IX. Future Work

Although a working design was calibrated and tested, future improvements could be made. Since the data-sheet was absent for the current sensor, calculations based on loose assumptions and estimates were used to convert voltage to pressure for the results obtained in this paper. An appropriate piezoelectric sensor with a complete data-sheet would result in an accurate conversion from voltage to pressure, which would lead to more scientifically accurate blood pressure readings. Moreover, the current circuit could be improved to increase precision and amplitude while decreasing the noise around the signals.

In order to obtain the most accurate data non-invasively, knowing the exact location of the artery is necessary. The current system employs a guess-and-check method in which the operator manually locates the pulse and places the sensor and actuator in the general area. The use of ultrasound could provide visual assistance in locating the artery. This becomes extremely important when placing the actuator, since it is quite easy to miss the artery and get inaccurate data. Furthermore, the ultrasound will show how much the artery is compressed. Currently, the operator subjectively determines when the artery is compressed halfway by estimating the maximum amplitude of the systolic peak on an oscilloscope. By implementing these changes, the time to take the measurements could drastically be reduced, making this system a more attractive clinical option.

The solenoid is currently powered with a DC power supply and an operator performs the actuation manually. The actuation is dependent on the response of the operator to turn the power supply on and off. In order to ensure consistent impulse durations, a microcontroller can be programmed to power the solenoid with a specific voltage for a set period of time. This voltage could be adjusted more easily with a controller based system to provide more adjustability for arterial depth and other factors. In addition, the timing of actuation can be automated by computer analysis of an electrocardiogram. A program could transmit a signal to actuate the solenoid at a specific time after the systole. Not only would making the actuation process automated

improve the quality of data, but it would also make the system quicker and easier to operate.

The data obtained for arterial stiffness is not well understood at this point. Further research into the significance of the waveforms is necessary to convert the relatively qualitative data to quantitative data. However, it is quite likely that this would be made easier by implementing many of the ideas already mentioned, making the data more consistent and easier to interpret. Examples of further data analysis possibilities include integrating the area under the artery response curve and modeling the response of the artery as a second order ordinary differential equation. These methods could be used to potentially convert the disruption in the waveform into quantitative data about the elastic properties of the artery.

Once the design is optimized, approval from the Institutional Review Board is necessary to test the design on human subjects. Subjects ranging in age, sex, race, and body composition, among others, will be tested. The data gained from different demographic groups and a wider sample size would help to verify the validity and accuracy of the system before clinical implementation.

X. References

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XI. Appendix A: Product Design Specifications

Arterial Actuator: Product Design Specifications
December 8, 2010
Client: Prof. John Webster
Team: Nick Thate, Evan Flink, Clara Chow, Henry Hu
Advisor: Amit Nimunkar

Problem Statement

Cardiovascular disease is one of the top killers in today's society. Blood pressure and arterial stiffness are indicator s of cardiovascular health. Currently, blood pressure is measured via sphygmomanometry and arterial stiffness via arterial tonometry. Although effective, the speed and accuracy of these methods can be improved. The goal is to design a system, comprised of a piezoelectric pressure sensor, an actuator, and a stabilizing structure, that can quantitatively measure blood pressure and arterial stiffness on a single artery.

Client Requirements

- Increased speed from current measurement methods
- Effective measurement of blood pressure and arterial stiffness
- System to incorporate all components
- Comparison to standard measurements

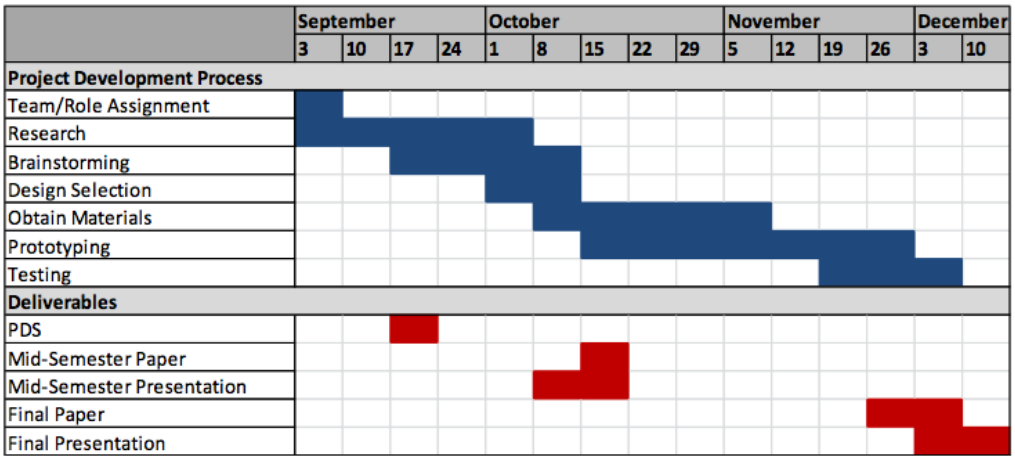
Design Requirements

1. Physical and Operational Characteristics
    a. *Performance requirements:* Measures blood pressure and arterial stiffness non-invasively on a single artery; must acquire data more quickly than current methods
    b. *Safety*: Cannot cause pain to patient, contain hazardous or sharp materials, or cause health problems
    c. *Accuracy and reliability:* Blood pressure and arterial stiffness data must be as accurate as data obtained from current methods; data must be reproducible and consistent between trials
    d. *Life in Service/Shelf Life:* The device should last at least 10 years
    e. *Operating Environment:* Indoor clinic, hospital, or laboratory
    f. *Ergonomics:* The system is modified to place the user in the most comfortable position
    g. *Size:* Should be able to fit in a cupboard or on a bedside table
    h. *Weight:* Must be light enough to be handled by an average person
    i. *Materials:* Piezoelectric sensor, push/pull solenoid, PVC pipe, wood base, light fixture whip, blood pressure cuff, aluminum block, circuitry
    j. *Aesthetics, appearance, and finish:* Should be presentable in a hospital setting

2. Production Characteristics
    a. *Quantity:* One prototype
    b. *Target Product Cost:* System less than $100

3. Miscellaneous
    a. *Standards and Specifications:* Meets FDA standards for a Class II medical device
    b. *Customer:* Hospitals, clinics, and laboratories
    c. *Patient-related concerns:* Cannot pose health or safety risks
    d. *Competition:* Standard sphygmomanometry and arterial tonometry devices, ambulatory blood pressure monitoring devices, pulse wave velocity blood pressure sensors

XII. Appendix B: Semester Schedule/Timeline



XIII. Appendix C: Expenses

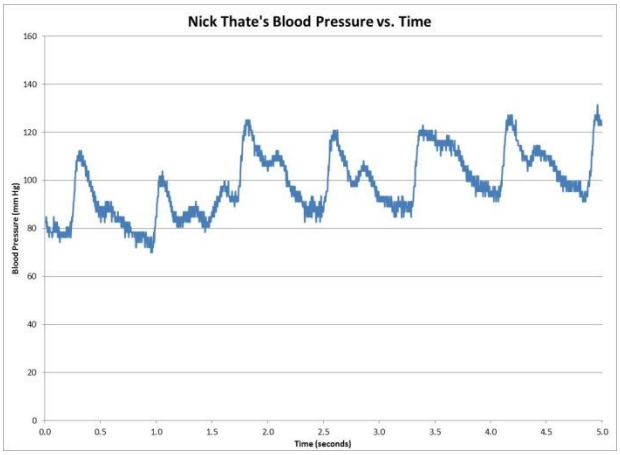
Product	Cost
Manual inflate blood pressure kit	$18.99
Linear push/pull solenoid	$32.96
PVC for arm stabilizer	$9.24
Miscellaneous construction materials	$22.68
TOTAL	$83.87

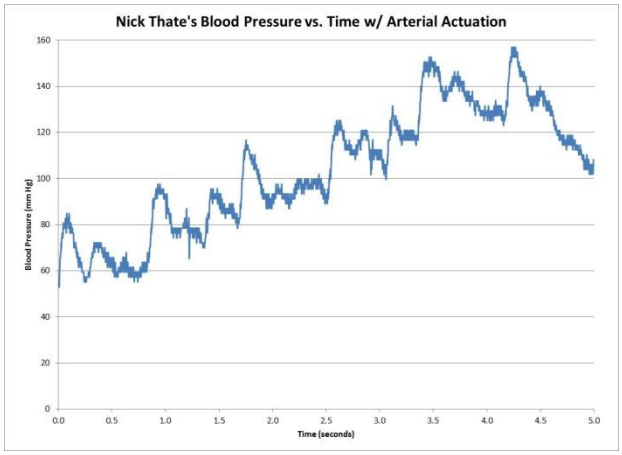
* Note: All dollar amounts listed above are before taxes and/or shipping *

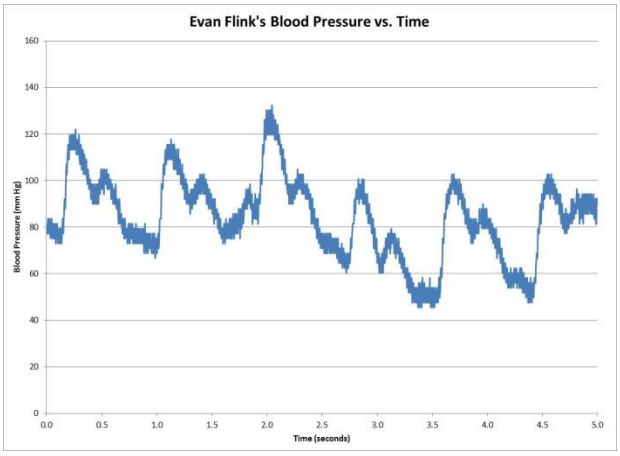
XIV. Appendix D: Net List for Wire-Wrapping Circuit

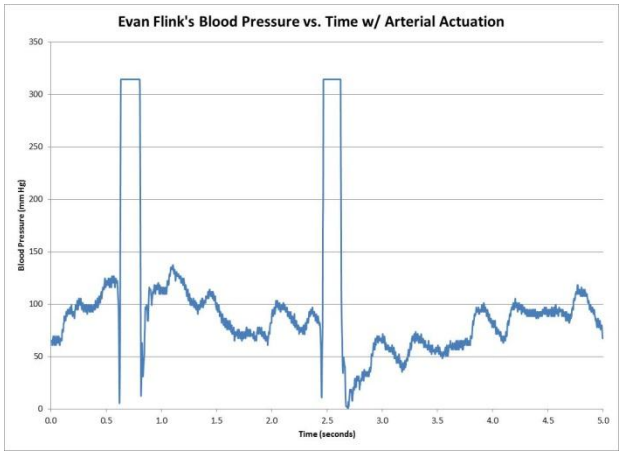
C2	GND
C2	Cdd(+8V)
Vdd	IC1(7)
C3	GND
C3	Vss(-8V)
Vss	IC1(4)
C4	GND
C4	Vref(+2.5V)
Vref	IC1(5)
R1	IC1(3)
R1	IC1(2)
C1	IC1(3)
C1	IC1(2)
Sensor(1)	IC1(3)
Sensor(2)	IC1(2)
R9	GND
R9	IC1(2)
R8	IC1(3)
R8	GND
R10	IC1(1)
R10	IC1(8)
IC1(6)	OUTPUT

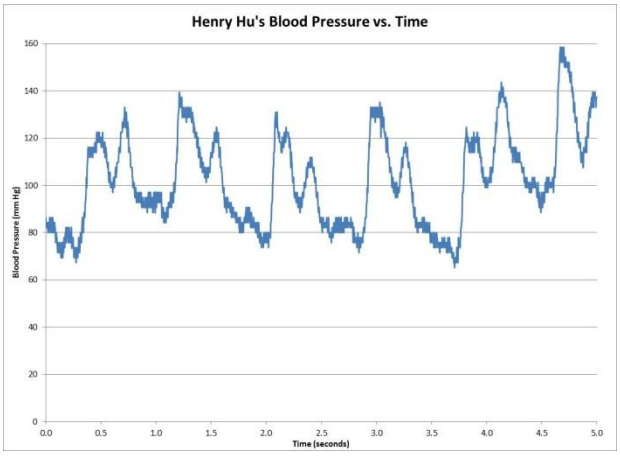
XV. Appendix E: Blood Pressure Waveforms

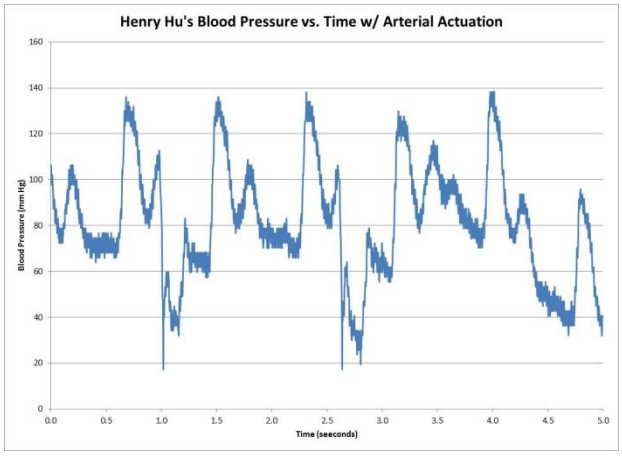


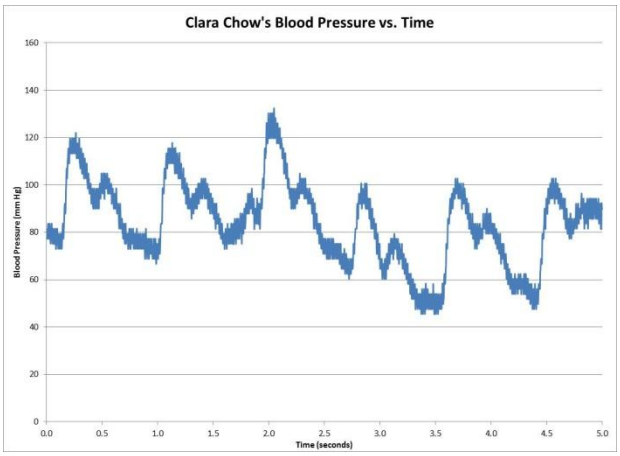


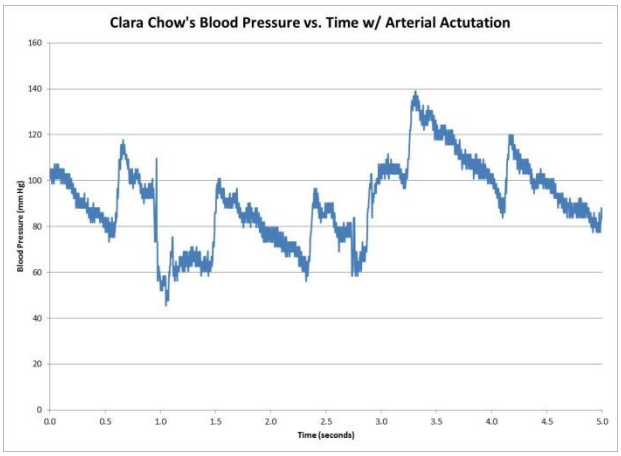


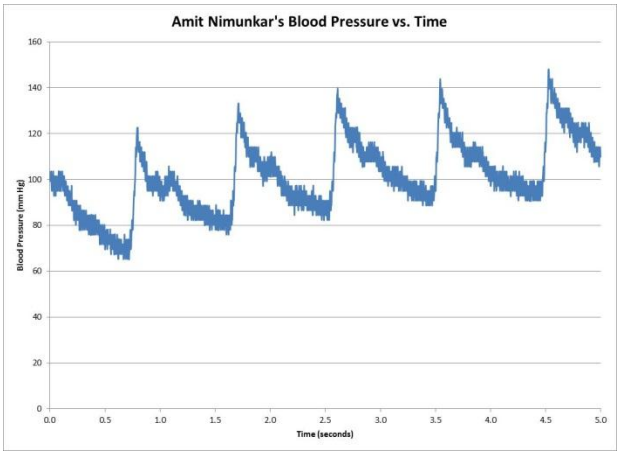


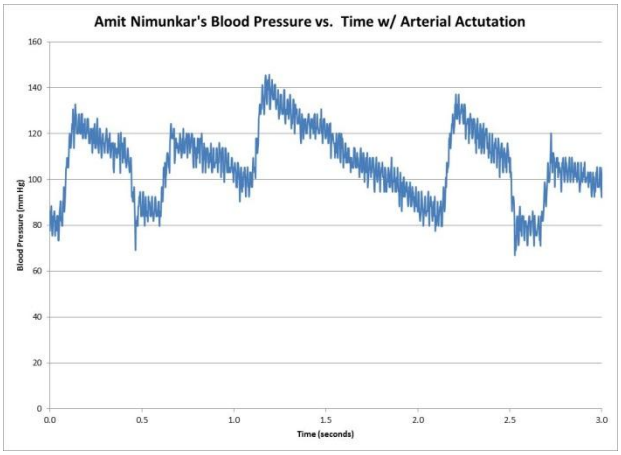





Clara Chow's Blood Pressure vs. Time
Blood Pressure (mm Hg)
Time (seconds)


Clara Chow's Blood Pressure vs. Time w/ Arterial Actutation
Blood Pressure (mm Hg)
Time (seconds)


Amit Nimunkar's Blood Pressure vs. Time
Blood Pressure (mm Hg)
Time (seconds)


Amit Nimunkar's Blood Pressure vs. Time w/ Arterial Actutation
Blood Pressure (mm Hg)
Time (seconds)

XVI. Appendix F: Testing Data

Person	Nick	Evan	Henry	Clara	Amit	Average
Recorded Systolic	120.5981	112.9811	140.1314	112.9811	137.4574	**121.6729**
Recorded Diastolic	81.8240	61.0297	75.1037	61.0297	80.6291	**69.7468**
Recorded Pulse Pres.	38.7741	51.9514	65.0277	51.9514	56.8283	**51.9261**
Actual Systolic	112	116	125	104	120	**114**
Actual Diastolic	74	82	80	70	80	**77**
Actual Pulse Pres.	38	34	45	34	40	**38**
Pulse Rate	46.3250	51.8100	47.7900	51.8100	55.9875	**49.4338**
Systolic Dev.	8.5981	-3.0189	15.1314	8.9811	17.4574	**10.6374**
Diastolic Dev.	7.8240	-20.9703	-4.8963	-8.9703	0.6291	**8.6580**
Pulse Pres. Dev.	0.7741	17.9514	20.0277	17.9514	16.8283	**14.7066**
Avg. Delay	0.0062	0.0050	0.0038	0.0012	0.0038	**0.0041**
Avg. Signal Length	0.1663	0.1538	0.1875	0.1388	0.1700	**0.1616**
Avg. Resp. Length	0.0563	0.2888	0.2363	0.2375	0.2538	**0.2047**
Avg. Resp. Deflection	18.0243	311.1749	92.2407	56.1921	54.0716	**119.4080**