Guardian A Wearable Medical Alert System

A Design Report

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Abstract

Master of Electrical Engineering Program Cornell University Design Project Report

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Abstract:

Guardian is a medical alert system in the form of a wearable device that monitors rapid shifts in blood pressure and acceleration to automatically detect when a user has fallen. When a fall is detected, the device will connect to a mobile device to call for emergency help. The focus of the design project is to develop the blood pressure subsystem, which is what differentiates Guardian from existing alert systems for seniors. This subsystem is comprised of two pulse plethysmography units that connect with an ATMega1284P microcontroller to continuously measure user blood pressure based on pulse wave transit time. The microcontroller measures the pulse wave transit time between the two pulse plethysmographs and applies an experimentally-derived regression model to calculate blood pressure. Rapid shifts in pressure indicate shifts in body orientation, suggesting the user has fallen. The goal of this project is to develop this optical blood pressure monitoring system and to see if it is a plausible addition to existing wearable medical alert systems.

Executive Summary

This design project was pursued to investigate more reliable means for fall detection among seniors. The goal of the project is to create an optical blood pressure monitoring system through a microcontroller based design. This system is also assessed to see if it could be beneficial to existing medical alert systems.

The system should be noninvasive, robust, and reliable to be used commercially.

The complete design uses two pulse plethysmographs placed at two different locations along the anterior side of a user's forearm, spaced five inches apart. Each pulse plethysmograph is comprised of a 940nm wavelength infrared LED and an OPT101 monolithic photodiode. When a pulse wave passes under a photodiode, a voltage peak is generated. The analog voltage outputs from each photodiode feed into analog-to-digital converter pins on an ATMega1284P microcontroller.

The microcontroller implements peak detection by checking for values above a defined threshold. After defining where the peaks are, the microcontroller measures the amount of time it takes for a pulse wave to travel between the two points on the user's arm. This time measurement is then used in a regression model to convert the pulse wave travel time to a blood pressure reading.

Overall, the cost of materials was less than \$22 and the system showed potential results for optical blood pressure monitoring on a user's forearm.

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Introduction

Background

Older adults are at a higher risk for many types of injuries that can lead to death or disability. According to the Centers for Disease Control and Prevention, falls are the leading cause of injuries and death for Americans aged 65 years and older.¹ Each year, about 35% to 40% of adults 65 and older fall at least once.¹ With a population of roughly 43 million people in this age demographic, this equates to more than 17 million falls a year.²

Medical alert systems were first introduced in the 1970s as basic push-button devices worn around the neck.³ These devices called for help by transmitting a signal a base station that was connected to a home phone line that could then alert a call-center operator. However, these alert systems require the user to press a button when they have fallen. Only when the user has pressed this button will emergency services be contacted.^{4,5} There are often situations where the user is incapacitated and unable to reach for this button when they have fallen. This has led to "Fall Detection" technology used in newer medical alert systems.⁶

This technology uses accelerometers and gyroscopes to analyze changes in a body's position for fall detection. While this automated fall detection seems to be an ideal solution, this detection has limited functionality and issues with reliability.⁷ Fall detection devices must be sensitive enough to separate everyday activities from falls. While newer medical alert systems have implemented rough algorithms for this detection, the technology must still be refined to catch edge cases, such as falls caused by dizziness, users sliding down from chairs to the floor, or when the user grips an object during the fall. False alarms could also be triggered by rapid movements by the user, inconveniencing both the user and the medical system dispatchers.

The proposed system incorporates continuous blood pressure monitoring as a means to supplement accelerometer and gyroscope readings for more robust fall detection. According to the American Heart Association, a sudden decline in blood pressure can cause dizziness or fainting.⁸ Changes in blood pressure can also indicate changes in body orientation. Using this information, a more robust medical alert system can be implemented.

Papers have shown that systolic blood pressure can be measured using pulse wave transit time (PWTT), which is the time it takes for blood to reach an extremity from the heart.⁹ The idea is that blood pressure affects the amount of time it takes for blood to travel through arteries. By creating a linear regression model to establish the relationship between PWTT and measured blood pressure, a system can be calibrated to measure blood pressure by applying pulse plethysmography at two different points on a user's body.

Existing Products

The creators of current medical alert systems emphasize that their products will aid users who might have fallen and cannot stand back up. The detection portion of these existing system designs can be divided into two categories:

- 1. simple push button
- 2. push button with motion-sensitive fall detection technology

Systems such as Life Alert and Bay Alarm Medical require the user to press a button to indicate that they have fallen.^{4,5} However, the user is sometimes incapacitated when they have fallen and will not be able to reach for the button.

Improved systems, such as Alert-1 include fall detection via readings from a gyroscope and accelerometer.⁶ While this design is an improvement on the simple push-button design, it has limited functionality and has issues with reliability.

To resolve the issues with present day solutions, the inclusion of a blood pressure monitoring subsystem is explored as a supplement to the fall detection technology.

Design Problem

The goal of this project is to design a proof of concept system for a continuous optical blood pressure monitoring system that can supplement existing fall detection methods. It is only one component of a complete medical alert system, named Guardian. Guardian is a wearable medical alert system that monitors rapid shifts in blood pressure and acceleration to automatically detect when a user has fallen. When a fall is detected, the device will connect to a mobile device to call for emergency help.

The pulse wave travel times measured by the blood pressure subsystem should be compared to blood pressure as captured by a commercially available blood pressure cuff to create a regression model, establishing a relationship between pulse wave travel time and blood pressure.

This deliverable will show how blood pressure changes can indicate changes in body orientation. Future work will be necessary to incorporate the accelerometer, gyroscope, and alert system to create a complete system. The complete system will be particularly useful for seniors who might be living alone and have a high risk of falling and becoming unconscious. By having an automated alert system, seniors will be able to live safely at home with a feeling of security that emergency help will come if needed.

System Requirements

Since the product is designed for seniors, a main priority is to keep the device easy-to-use with minimal maintenance. The system requirements are as following:

- the system shall be low cost (< \$50)
- the system shall be compact and worn on the forearm
- the system shall be useable by multiple users

A main constraint for this project is that it is to be executed by one person within three credits. Because of this, many tasks were re-scoped – unlike the original plan, multiple design iterations were not possible and data collection to create the regression model was limited.

Range of Solutions

A range of solutions were explored before selecting the current design for blood pressure monitoring. These solutions included different methods to measure pulse wave travel time.

The pulse wave travel time could have been calculated by the time difference between an electrocardiogram (ECG) and a pulse plethysmograph on the wrist. This method requires electrodes on both sides of the heart for the ECG. Another option was to calculate pulse wave travel time between the index fingers or wrists of both arms, using bluetooth modules to wirelessly communicate between the disjoint areas. While both these solutions could have rendered more accurate blood pressure readings, both options are bulky and would be inconvenient to use on a daily basis.

The primary objective for this subsystem is to detect rapid changes in blood pressure to aid in fall detection. Thus, the precision of the blood pressure reading is not as important as detecting the changing blood pressure. For the purposes of this subsystem, it is a fair tradeoff to aim for a compact device with less accuracy. To achieve a more compact form factor, the current design implements a sleeve form factor with embedded pulse plethysomgraphy sensors spaced five inches apart. This sleeve will measure pulse wave transit time along a user's forearm. This design, which is confined to a single forearm, is less intrusive to a user's daily life.

System Design

The full fall detection system is divided into multiple subsystems. While the full system is defined in Figure 1, the focus for this project is solely the blood pressure subsystem.



Figure 1. Guardian high level system design

A fall is defined when the accelerometer and gyroscope first detect a rapid change in acceleration or when the blood pressure subsystem detects a rapid change in blood pressure. When one subsystem is triggered, the microcontroller unit senses this change and checks the other subsystem to confirm if a fall was also detected. If a fall is confirmed by both the fall and blood pressure subsystems, the microcontroller provides an audio and visual indication that a fall has been detected. If the user does not disable the system within five seconds, the microcontroller will automatically connect to a mobile device via a bluetooth module to contact emergency services.

The following table is a behavioral description template (BDT) for the device's primary use case: when the user wears the device for fall detection. The table demonstrates in detail how the device subsystems interact with one another and the user

Table 1. Behavioral description template for "user wears device for fall detection" use case

User	Gyroscope/Acc elerometer Subsystem	Blood Pressure Subsystem	Microcontroller	Alert Subsystem
wears Guardian on forearm				
falls	fall detected	rapid shift in blood pressure		
			measures changing pulse wave travel time confirms fall detected	
			5 second time delay for user to turn off system if false positive	
			sends bluetooth transmission to trigger alert system	
				system contacts call center operator for emergency services

Initial Condition: device is on and at rest

The following diagram shows the form factor of the full Guardian system on a user's arm.



Figure 2. Guardian system form factor

Theory of Operation

Pulse Wave Transit Time

Pulse wave transit time (PWTT) is the amount of time it takes for a pulse pressure wave to propagate through a length of the arterial tree.¹⁰ When the heart pumps blood from the left ventricle into the aorta, it creates a pressure wave that travels along the arteries ahead of the pumped blood itself. This pressure wave travels at a speed of a few meters per second.

While PWTT is typically measured using readings from an electrocardiogram (ECG) and a peripheral pulse wave via a pulse plethysmograph, it can also be measured between two points on a user's body with two pulse plethysmographs.



Figure 3. Illustration of PWTT between ECG and pulse plethysmograph

Because the speed of this pressure wave depends on the tension of the arterial walls, PWTT correlates with a user's blood pressure. The higher the user's blood pressure, the faster the pressure wave travels, resulting in a shorter PWTT. The lower the user's blood pressure, the slower the pressure wave travels, resulting in a longer PWTT. Using this knowledge, a linear regression model can be created for a correlation between PWTT and blood pressure.

Pulse Plethysmography

Pulse plethysmography is a method used to measure a user's heart rate. It is a commonly used method because it is noninvasive, inexpensive to implement, and can be very compact.

When the heart beats, a pressure wave moves away from the heart along the. This pressure wave creates an increase in blood volume in tissue wherever the wave is.

Using an infrared LED and photodiode, this volumetric change in blood from the pressure wave can be detected. The infrared light is directed into a region of tissue with a paired photodiode pointed next to that region. Infrared light is used because it is relatively unaffected by changes in arterial oxygen saturation. Some of the light is absorbed by the tissue and deoxyhemoglobin while the rest of the light is reflected back. The photodiode detects the amount of infrared light that is reflected back from the tissue and outputs an analog voltage that linearly corresponds to the amount of infrared light detected. When the pulse wave reaches the region that the LED and photodiode are pointed at, the reflectivity of the tissue increases, causing an increase in voltage output from the photodiode. The composition of surrounding tissues remains the same as blood volume changes, which can add a DC component to the photodiode output.



Figure 4. Principle of reflectance pulse plethysmography, illustrating the optical sensor and the different layers in the skin¹¹

Peaks are detected in this reflectivity measurement – when the pulse wave is below the photodetector, a peak will occur. Thus, using pulse plethysomgraphy at different locations on the user's body can help monitor the location and speed of the pulse wave.

Background Math

Moens-Korteweg equation is describes the relation between blood pressure and pulse wave velocity (PWV, which is inversely correlated to PWTT).¹² It assumes that the pulse wave velocity in a short elastic vessel is obtained from its geometric and elastic properties as defined in the following equation:

(1)
$$PWV = \frac{distance}{PWTT} = \sqrt{\frac{Eh}{\rho 2r}}$$

where E = elasticity modulus of the vessel wall, h = thickness of vessel wall, $\rho =$ density of blood, r = radius of the vessel. Blood pressure and PWV are then correlated using the Hughes equation:

(2)
$$E = E_0 e^{\alpha P}$$

where $\alpha \approx 0.017 \text{ mmHg}^{-1}$ and P = mean arterial pressure. Along, each of the parameters in both equations are difficult to measure. However, calibration functions can be derived by combining these equations to translate PTT to blood pressure, assuming constant vessel thickness and radius. After combining the two equations, the following relationship is derived:

$$Blood Pressure = A \ln PTT + B$$

where A and B are constant factors that need to be experimentally derived. These constants can be derived by simultaneously measuring PWTT with the designed system and measuring blood pressure with a commercial blood pressure cuff. After plotting this data on the same set of axes, a regression model of the form of equation 3 can be fit to the data. This derived regression model can then be used to determine the relationship between blood pressure and PWTT.

Blood Pressure Subsystem High Level Design

The blood pressure subsystem uses two pulse plethysmographs placed at two different locations along the anterior side of a user's forearm, spaced five inches apart. One pulse plethysmograph is positioned on the anterior side of the user's forearm closer to the wrist while the other plethysmograph is positioned on the anterior side closer to the antecubital fossa (the area on the arm opposite the elbow). Each pulse plethysmograph is comprised of a 940nm wavelength infrared LED and an OPT101 monolithic photodiode. The infrared light is pointed toward the anterior side of the user's forearm next to the OPT101 which measures the amount of infrared light reflectance from the user. When a pulse wave reaches the tissue beneath the photodiode, the increased blood saturation increases the amount of light that is reflected to the photodiode, creating an analog voltage peak with the pulse wave. The analog voltage outputs from each photodiode feed into analog-to-digital converter (ADC) pins on an ATMega1284P microcontroller.

The microcontroller implements peak detection by checking for values above a defined threshold. After defining where the peaks are, the microcontroller measures the amount of time it takes for a pulse wave to travel between the two points on the user's arm. This time measurement is then used in a regression model to convert the pulse wave travel time to a blood pressure reading.

Blood Pressure Subsystem Hardware Design

In the blood pressure subsystem, the main components are a sensing subsystem, a filtering and amplification subsystem, and a microcontroller unit. While the final form factor of the Guardian system will be a sleeve worn on the forearm, the current prototype resides on a breadboard – where the user lays his or her arm above the sensors. For a full circuit schematic, refer to the appendix.



Figure 5. Breadboard prototype layout

The main sensing element for each pulse plethysmograph is the OPT101 sensor, which includes a linear light intensity to voltage conversion and a unity gain buffer. A single infrared LED at 940nm wavelength is used in each pulse plethysmograph to inject light into the region the OPT101 is monitoring. The signal from each sensor is buffered, filtered (both actively and passively), and amplified with two gain stages of 50 each before the signals are passed into ADC pins on the microcontroller unit.

Sensing Subsystem

Each sensing subsystem is comprised of a LTE4208 infrared LED at 940nm wavelength and an OPT101 monolithic photodiode. This particular photodiode was selected for this design because it includes a photodiode and a transimpedance amplifier to output voltage linearly with photocurrent.

The configuration of the LED and photodiode have a significant impact on the results. In this setup, the LED is positioned next to the photodiode, bent at an angle of about 120 degrees to direct its light toward the sensor. Both the LED and photodiode point toward the user's forearm.



Figure 6. User arm over photodiodes on breadboard

Filtering and Amplification Subsystem

Three types of filters are used in this design. To remove the DC offset from amplification, the signal from the OPT101 is first passed through a passive low pass filter with a cutoff frequency of 10 Hz before the input to the first amplification stage. An active bandpass filter/amplification stage as well as several passive low pass filters are then used to remove 100 Hz noise from overhead lights. The bandpass filter only allows frequencies between 1 Hz and 8 Hz to pass through. Typical human heart rates range between 30 to 180 beats per minute – the 1 Hz high pass removes low frequency noise from user movements while the 8 Hz low pass removes high frequency noise from ambient light and power lines.

Two gain stages with 30 gain each are used to increase the input signal to the microcontroller for easier peak detection. The gain stage is split into 2 op amps because too much gain in a single stage saturated the signal.

Microcontroller

The microcontroller used for the Guardian system is the ATmega1284P mounted on a custom printed circuit board. The design uses an ATmega1284P for its wide range of peripherals and to accommodate future integration of the "fall" detection subsystem. The blood pressure monitoring system uses two ADC input pins and a timer on the microcontroller. Future design iterations could implement the low-power mode of the microcontroller or use a lower powered CPU, such as the ATTiny85.



Figure 7. ATMega1284P microcontroller on custom PCB

Hardware Testing

To verify proper functionality of the circuit, each pulse plethysmograph circuit was systematically tested. Testing was performed on each individual filter and gain stage within the circuit by using a function generator as an ideal source. After ensuring each stage behaved as expected individually, stages were incorporated one by one to ensure proper hardware functionality.

Blood Pressure Subsystem Software Design

The main tasks of the microcontroller software include acquiring data from the photodiodes via ADC pins A.0 and A.1, processing the acquired data to extract PWTT measurements in real time, and converting the PWTT measurements into blood pressure readings. While this software was written, it was not implemented in conjunction with the hardware due to reliability issues in the hardware.

Analog-to-Digital Converter

The ADC is initialized using VCC as the voltage reference. The ADC value is then left aligned in the data registers to for 8-bit measurements rather than 10-bit. After these initializations, the ADC is enabled with a prescaler of 1.

The analog signals from each pulse plethysmograph are passed into ADC pins A.0 and A.1 after going through filtering and amplification. These signals are converted into digital values ranging from 0 to 255.

Peak Detection

Peaks for the pulse wave are currently detected in software using thresholding. If the current ADC value is above a predetermined value, the system recognizes the data to be a peak. This simplified method is susceptible to noise and drifting analog signals due to changes in body orientation or LED positioning. Thus, peak detection should be performed by saving a buffer of data from the last ten seconds.

The data should first be checked to see if it is valid. To do this, the number of detected major peaks in this buffer should be checked. The standard deviation of time between each peak should be used to determine if the signal is periodic and the heart rate in beats per minute (bpm) should be checked to see if it falls within typical human bpm rates (30-180 bpm). After a valid signal is established, the DC component of the signal should be removed by taking a fast fourier transform of the signal and removing the zero frequency component. An inverse fast fourier transform should then be performed. Peaks that occurred due to noise should then be removed by rejecting all peaks less than 25% of the amplitude of the maximum peak in the buffer. This creates a dynamic threshold for peak detection that constantly changes with the current data, which is more accurate than peak detection by simple thresholding.

Timing

Timer 0 is set up as a 1 millisecond time base for the system. This is done by enabling Timer 0 compare match ISR and enabling the clear-on-match. In the Timer 0 ISR, the subroutine continuously increments the counter time1.

When a peak is detected for the pulse plethysmograph closer to the elbow, the time variable is set to zero. This variable is incremented in the ISR for every passing millisecond. When the peak is detected for the pulse plethysmograph closer to the wrist, the milliseconds that have passed is recorded. This value is the PWTT that can then be used to calculate the user's blood pressure.

Hardware and Software Tradeoffs

There were a couple of signal processing components that could have been performed in either hardware or software. In general, anything that could reasonably be performed in software was calculated in software to minimize the size of the device.

For example peak detection was performed in software to minimize the amount of hardware used. In hardware, this detection could implemented using a comparator.

On the contrary, the signal filtering was performed in hardware rather than in software. To perform digital filtering, data would have to be stored in memory. The ATMega1284P has 16kB of RAM, which could limit the processing capabilities of the microcontroller which could slow down the software. With such a short distance between the two plethysmographs, it was more important to keep the software lean and quick. Thus, analog filtering was used.

Results

The system was tested on a single user with the following characteristics: female, age 22, height 5'7". Using the designed blood pressure subsystem, PWTT's were measured in various user orientations: standing, sitting, and supine positions. Five data points were taken for each user orientation. PWTT's for this plot were measured using a Tektronix oscilloscope and were compared to blood pressure readings measured by an Omron Healthcare's digital wrist blood pressure monitor model HEM-608.

Position	PWTT [ms]	SBP [mm Hg]
sitting	100	113
	55	122
	73	116
	82	110
	90	110
supine	85	100
	62	109
	94	108
	82	101
	67	103
standing	70	119
	80	115
	70	114
	72	110
	85	116

Table 2. PWTT and systolic blood pressure (SBP) measurements for sitting, supine, and standing orientations

The data in Table 2 show a higher systolic blood pressure reading from the commercial blood pressure cuff when standing and a lower systolic blood pressure reading when laying down. This trend supports the idea that changes in blood pressure can indicate changes in body orientation which could be useful in detecting when a user falls. However, the data has a range of 22 mmHg which is a narrow window. More data should be collected to develop a stronger relationship between PWTT and blood pressure.



Figure 8. Regression model relating systolic blood pressure to pulse wave travel time

The best fit regression model for the data was: blood pressure = $-13.0 \ln(PWTT) + 167.5$ with an R² value of 0.109, indicating poor correlation between the two variables. In Proenca et al¹², a best equation of blood pressure = $-92 \ln(PWTT) - 29$ was derived. While these derived constants are very different from the values derived from Figure 8, Proenca et al¹² have high variability in their values with a +/- factors of up to 333.

Speed of Execution

If the software were implemented with the hardware, the speed of execution would be limited by ADC conversion rate, which for an ATMega1284P is about 300µs. This is fast enough for the system to measure PWTT readings since travel times are on the scale of milliseconds.

Accuracy

The data collected for the regression model was very limited. The model was based on fifteen measurements taken from one female user with a defined height and age. These measurements were only taken in three positions: sitting, supine, and standing. Because of this, the range for measured systolic blood pressure is very small – a window of only 22 mm Hg. This is a small enough range and sample size of data that the accuracy of the collected data is uncertain. This regression model could be improved by increasing the range for systolic blood pressure by elevating blood pressure through exercise or lowering blood pressure through squeezing the brachial artery. The regression model could also be improved by collecting data from more users with greater diversity in demographics.

The distance between the two pulse plethysmographs is a very short distance of five inches. PWTT is typically measured between a user's heart and finger. This shorter distance for the pulse wave to travel can amplify the effects of any confounding components that can decrease the system accuracy.

Another component that has a significant effect on the accuracy of the blood pressure reading is the positioning of the sensors and infrared LED's on the user's arm. With flexible leads, the positions of these components can change each time the system is used. Thus, the collected PWTT readings might not have been based on the same positioning of LED's. On top of this, the reported testing results were performed in a stationary setting. Realistically, a user would move his or her arms while wearing the device. This arm movement could affect the system readings and the blood pressure itself. This could result in movement artifacts that could cause false peak detection and decreased accuracy. The amount of tension in the muscle under the photodiode can also affect the reflectance of infrared light.

Overall, there are multiple causes for inaccurate readings. The best way to overcome these factors in the given system design is to increase the amount of data collection to decrease the impact of inaccurate readings.

Safety

The safety concerns for this project are minimal due to the noninvasive nature of optical methods. The most dangerous components within the design are the low power infrared LED's. According to the International Commission on Non-Ionizing Radiation Protection (ICNIRP) standards on infrared radiation, direct eye exposure to small infrared sources for more than 0.25 seconds at a time should be limited to 1.9E07 W*m⁻²*sr⁻¹ to prevent thermal injury in the cornea.¹³ Users will be warned against looking directly into the infrared LED's. The LED's will also be embedded in a sleeve, so users will not be looking at the LED's directly.

To prevent thermal injury to the skin, the heat should be limited $2*t^{3/4} J*cm^{-2}$. Assuming the cross sectional area of skin exposure to the infrared LED is about $1cm^2$ and that 80mW of power is transmitted from the infrared LED with ideal transmission over thirty seconds, heat should be limited to a maximum of 2.4 J/cm².

Interference with Other Designs

The blood pressure subsystem itself does not have any components that could interfere with other devices. However, the addition of the bluetooth module for connecting to mobile devices has potential to interfere with other devices. This would be unlikely – as long as the integrated bluetooth module conforms to standards outlined by the Bluetooth Special Interests Group, it should not interfere with other devices.

Usability

Since the device is meant to be used by seniors, usability is a key component. The prototype in its current form factor on a breadboard is not portable for a user. However, the complete Guardian design will be portable and easy to use. With an elastic sleeve design that contains all subsystems on a single forearm, the device should be easily portable and should not interfere with a user's daily habits.

If enough user testing is performed to create a robust platform of regression models that can be tailored to the user's age, height, weight, and gender, the device should be able to be used for all users.

The concerns for the device usability are potential false positives that could be triggered by the user swinging his or her arms. Another concern is system sensitivity to the infrared LED orientation in comparison to the photodiode. On the breadboard, it was very difficult to consistently place the LED's in an orientation that could produce usable data. Thus, a different form factor that is worn on the torso should be considered. The torso is likely to have less movements in comparison to a user's arm.

Conclusions

This optical blood pressure monitoring system was built to supplement existing methods of fall detection using gyroscopes and accelerometers for fall detection. To enhance existing methods,

the blood pressure monitoring system must first be reliable on its own. The project shows potential to be integrated in future systems for increased fall detection reliability. However, further testing needs to be conducted. Regression models need to developed for various users based on age, height, weight, and gender for the product to be applicable for a wide range of users. This will require abundant data collection.

Although the current design was favored over alternate designs in the beginning of the project, different form factors should be explored for the design to be more robust. With the current design, the positioning of the infrared LED's and photodiodes can be inconsistent, which can have great impact on readings. Additionally, users tend to move their arms around even while stationary. This effect can cause false readings. A form factor on the torso area, which experiences less movement, should be explored.

Work Accomplished

The main tasks that were completed are summarized in Table 3. A prototype for continuous optical blood pressure monitoring was developed. An initial linear regression model was also developed to correlate blood pressure with pulse wave transit time as measured by the system.

	September C			Oct	October			November			December					
Task	1-Sep	8-Sep	15-Sep	22-Sep	29-Sep	6-Oct	13-Oct	20-Oct	27-Oct	3-Nov	10-Nov	17-Nov	24-Nov	1-Dec	8-Dec	15-Dec
Brainstorm project																
ideas																
Decide on project																
and write abstract																
Perform research on																
similar products																
Develop high level																
system design																
Write project																
proposal																
Build blood pressure																
subsystem prototype																
Develop linear																
regression																
Determine																
effectiveness of																
method																
Write final report																

Table 3. Project schedule

Future Work

The reliability of the blood pressure monitoring system could be enhanced by pursuing different form factors. One option is to measure pulse wave transit time by applying electrocardiography

and pulse plethysmography on a user's deltoid muscle, which has shown potential for providing ECG data. This form factor would be a sleeve worn on the user's upper arm as opposed to the user's lower arm. Another option would be to measure PWTT by applying electrocardiography and pulse plethysmography across a user's chest in the form factor of a band that could be worn under a user's shirt.

More importantly, to measure blood pressure more accurately, more data needs to be collected. This data needs to span a wider demographic of users in a greater variety of use cases. Because it is difficult to artificially decrease blood pressure, the use cases should aim to increase blood pressure to create a greater window of data for the linear regression model. Such use cases could include measuring PWTT after rigorous exercise or while squeezing the brachial artery.

Machine learning could also be applied to help reduce the amount of data collection required to develop robust regression models.

Standards

The C code embedded in the microcontroller complies with the C language standards set by the American National Standards Institute and the 1666-2011 IEEE Standard System Language.

Ethical Considerations

During the development process of this prototype design, the IEEE Code of Ethics was followed. Conscious design choices were made to ensure safety for the public and for myself while developing a quality product. While constructing the system, I constantly checked for short circuits, corrupt components, and made sure all electrical components were properly insulated.

Another aspect of ethics that was considered was the trust that seniors would have in this device. By advertising this device to be supplementary to existing methods, users would expect this device to perform more reliably. Thus, the final system design will include the same basic functionality as existing devices on the market, including a push button and accelerometer/gyroscope fall technology.

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 http://www.icnirp.org/cms/upload/publications/ICNIRPVisible Infrared2013.pdf>

Data Sheets

Infrared LED LTE-4208

http://people.ece.cornell.edu/land/courses/ece5030/labs/s2014/E4208irEmitter.pdf

Monolithic Photodiode OPT-101

http://www.ti.com/lit/ds/symlink/opt101.pdf

ATMega1284P Microcontroller

http://people.ece.cornell.edu/land/courses/ece4760/AtmelStuff/mega1284full.pdf

Omron Healthcare digital wrist blood pressure monitor model HEM-608 http://www.airforcemedicine.af.mil/shared/media/document/AFD-130404-090.pdf

Appendices

Cost and Budget

Part	Vender	Cost/Unit	Quantity	Total Cost
Infrared LED LTE-4208	Digikey	\$0.50	2	\$1.00
Monolithic Photodiode OPT-101	Digikey	\$6.37	2	\$12.74
LM358 Op Amp	Digikey	\$0.50	2	\$1.00
ATMega1284P	already owned	\$5.00	1	\$5.00
Capacitors	lab stock	\$0.20	10	\$2.00
Resistors	lab stock	\$0.00	12	\$0.06
				\$4400

Total Cost \$21.80

Circuit Schematic



Code Listing #include <util/delay.h> #include <stdio.h>

```
#define F_CPU
                     1600000UL //frequency of CPU crystal
                                    //peak detection threshold
#define PP1 thresh
                     100
#define PP2_thresh
                                    //peak detection threshold
                     150
                             -11.7
                                           //scaling factor for regression model
#define k1
#define k2
                             162.2
                                           //offset factor for regression model
                             time1: //time counter for PWTT
volatile unsigned char
volatile unsigned char
                             PWTT:
                                           //pulse wave transit time
volatile unsigned float
                             BP;
                                           //blood pressure
// timer0 overflow ISR - used to increment a timer, keep a millisecond // time base
ISR (TIMER0 COMPA vect)
{
//Increment the time
 time1++:
}
// Reads adc value at pin ch
// Code borrowed from ADC tutorial:
// http://maxembedded.com/2011/06/20/the-adc-of-the-avr/
// @param
                     which pin (which transceiver) to read: 0=PP1, 1=PP2
              ch
// @return
                     adc-converted reading
uint8_t adc_read(uint8_t ch)
ł
  // select the corresponding channel 0~7
  // ANDing with '7' will always keep the value
  // of 'ch' between 0 and 7
  ch \&= 0b00000111;
                               //AND operation with 7
  ADMUX = (ADMUX \& 0xF8)|ch; //clears the bottom 3 bits before ORing
  // start single conversion - write '1' to ADSC
  ADCSRA \models (1 \leq ADSC);
  // wait for conversion to complete
  // ADSC becomes '0' again
  // till then, run loop continuously
  while(ADCSRA & (1<<ADSC));
  return (ADCH);
}
// initialize adc
```

```
void adc_init()
{
       // AREF = AVcc
       ADMUX |= (1<<REFS0);
       ADMUX |= (1<<ADLAR); // left align ADC value in data registers
                     // (turn measurement from 10 bits to 8 bits)
       // ADC Enable and prescaler of 1
       ADCSRA = (1 << ADEN);
}
// Set up the MCU
void initialize(void)
{
 // set up timer 0 for 1 mSec timebase
 TIMSK0 = (1<<OCIE0A); //turn on timer 0 cmp match ISR
 OCR0A = 249;
                             //set the compare reg to 250 time ticks
 TCCR0A = (1<<WGM01); // turn on clear-on-match
 time 1 = 0;
                            // init the task timer
 DDRA &= \sim (0x03);
                             // set a.0 and a.1 to input
                             // crank up the ISRs
 sei();
}
// Check for peak in pulse pleth 1 (closer to elbow)
int checkPP1(void)
{
       if (adc_read(0) > PP1_thresh)
              return 1;
}
// Check for peak in pulse pleth 2 (closer to wrist)
void checkPP2(void)
{
       while (adc_read(1) < PP2_thresh); //wait until peak detected
       PWTT = time1;
}
int main(void)
ł
       initialize();
       adc_init();
       // main task scheduler loop
       while(1)
       {
```

```
if (checkPP1() == 1)
{
    time1 = 0;
    checkPP2();
    BP = A*log(PWTT)+B;
}
}
```