Wearable Optical Cardiovascular Biometric Tracking System

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Abstract – The objective of this project was to design and construct a wearable device with an optical sensor which interfaces with a personal device to provide cardiovascular biometrics to a user. The system is able to provide Heart Rate, Heart Rate Variation, Blood Oxygen Content, and Skin Temperature.
Index Terms – Photoplethysmogram, Biometrics, Wearable Electronics, Signal Processing, Photonics.

I. Introduction

The wearable optical biometric tracking system proposed by our group is a mobile friendly, and easy-to-use device that enables users to monitor their cardiovascular health in their day-to-day life. The proposed device will utilize PPG technology and interface with a user's personal device, such as their smartphone, to collect and store biometric data for an extended period of time such that a user can observe their progress with respect to their cardiovascular health with ease. With our device, we hope to enable users to take charge of their cardiovascular health and make improvements by providing them a method for benchmarking their cardiovascular performance and facilitating the access to health feedback.

The basic goals that our group has for the project include being able to measure and provide feedback on the 4 following biometrics:

1. Heart-rate
2. Heart-rate variability
3. Blood oxygenation (SpO2)
4. Skin temperature

All of the biometrics are tied to cardiovascular health as blood oxygen and thermoregulation issues are early indicators of serious health issues. Other basic goals that our group has include the ability to wirelessly communicate with a personal device to facilitate the system's operation, as well as constructing a highly efficient optical sensor that collects the PPG signal with the least amount of noise and errors possible.

Goals and Motivation. Mindful monitoring of one's cardiovascular health is a useful habit to establish in the modern day as cardiovascular disease (CVD) is the leading cause of death worldwide. Identifying an individual's personal cardiovascular biometrics, such as heart rate, is a good approach towards analyzing their risk for CVD, as well as monitoring progress towards better cardiovascular endurance. There currently exist a few different methods to extract cardiovascular biometrics, with one of the most popular being an EKG system, in which multiple electrodes generate an electric signal with embedded cardiovascular information. While this method is considered the gold standard by the medical community, it offers disadvantages with regard to personal health tracking as using EKG systems can be cumbersome to implement in day-to-day life due to the need for various probes for measurement. Optical sensors provide a significant advantage in this regard due their ability to be implemented in small dimensions that facilitate casual wearable designs. An optical technique that is becoming increasingly used in personal biometric tracking systems is the generation and processing of a PPG signal via an optical sensor system.

A PPG sensor system obtains information about a person's cardiovascular system by measuring the change in volumetric blood flow from blood vessels in the skin. Given that the PPG signal is derived from a person's blood flow, various biometrics that could not be extracted from techniques like an EKG are potentially possible given careful enough analysis of the PPG signal. It is not unusual to derive biometrics such as heart-rate, heart-rate variability, and blood oxygenation from the PPG signal. In addition to the previously stated biometrics, other critical health-related biometrics such as blood pressure (diastolic and systolic), blood glucose, and arterial stiffness can be derived from the PPG signal and are currently the focus of various biomedical engineering researchers. Research in this academic field has also focused on the reduction of errors that affect the signal quality of PPG signal as it is a prerequisite to accurate extraction of biometrics, although a noticeable amount of developments done in these studies has yet to be implemented in commercially available devices.

Due to the great potential that PPG technology offers in the future and the advantages it currently offers in mobile health monitoring, a large number of commercial PPG-based devices have been developed and sold in the health device industry. A push for more fitness-oriented PPG-based health tracking devices can be seen in the marketplace from such companies as Fitbit and Polar via their mobile-device compatible products.

Our group's motivations for this project are to develop a biometric tracking system that is convenient to use for the majority of people so that they can accurately know their cardiovascular health within their day-to-day life. An additional motivation of our group is to attempt to bridge the current gap between the academic developments of PPG system designs and the marketplace-friendly designs that are commonplace of commercial products. This second motivation is meant to assess the feasibility of optimized PPG system designs with consideration to cost, time and practicality.

Our group's basic project goals are to design and construct a non-invasive wrist-worn biometric monitoring device via the processing of optical bio-signals. The optical sensor will ideally be designed such that notable sources of errors with respect to signal acquisition are considered and mitigated as best as possible given present constraints. The proposed device will be able to interface wirelessly with a personal device via a health monitoring application that will store biometrics for an extended period. The long-term storing of biometric information enables a user to observe trends with respect to their biometrics to assess their cardiovascular health's improvements or deterioration. The principal biometrics that are to be collected for the basic design goals are the following: Heart rate, Heart Rate Variability (HRV), Blood oxygenation concentration (SpO2), and Skin temperature.

II. High-Level Overview

The planned system will incorporate the function of a designed device that uses LEDs and sensors to gather biometric data. It will then analyze the information gathered and use a communication protocol to push the results to another device for display and storage. The various components will be completed by each of the members of the team. Each member of the team will be responsible for the completion of at least one of these components from an administrative perspective but any or all of the team may end up working on any given component. Responsibility for each is labeled in the figure below

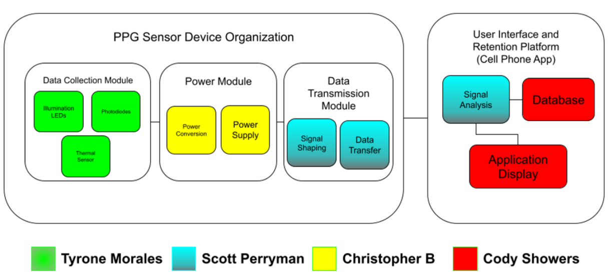

Fig. 1. Subsystem assignment diagram.

III. Hardware

Device Sensor. For the on-device sensor we utilized a PPG sensor and a optical thermal sensor to collect biometric data.

Photoplethysmography (PPG) is an opto-electronic technique that utilizes the temporally varying optical properties of skin arising from volumetric blood flow from micro-vascular tissues to measure cardiovascular information. PPG sensors are comprised at a fundamental level of simply an emitter and a photo-detector arranged in either a reflection or transmission configuration. The transmission configuration utilizes an emitter and detector on opposite sides of a limb, typically a finger, to generates a PPG signal from the collection of attenuated light as it passes through a subject's limb. The reflection configuration utilizes the emitter and detector on a singular side and therefore relies on the collection of diffusely reflected light from the blood vessels to generate a signal. For this product, our team decided to design our sensor to work in the reflective mode of operation in order to facilitate easy of use. figure

The designed PPG sensor utilized a SFH7016 multiwavelength LED chip and the SFH2202 photodiode which were both manufactured from OSRAM. The SFH7016 generates green (526 nm), red (660 nm), and near-infrared (950 nm) light with a linewidth of less than 50 nm. The SFH2202 was chosen due to it's complementary responsivity spectrum for the selected wavelengths, providing a responsivity of: 0.32 A/W for 525 nm, 0.473 A/W for 660 nm, and 0.67 A/W for 950 nm. These wavelengths were chosen due to the absorption spectra of particular chromophores present in skin that give ways to it's effective optical properties, as can be seen in Figure 1. Melanin and water induce unwanted high attenuation for wavelengths higher than 450 nm and lower than 1 µm, while different types of hemoglobin, our target chromophore, has a notable response at the chosen wavelengths. Of particular interest for SpO2 measurements is the difference in the absorption spectra of oxygenated and non-oxygenated blood. Given that for the chosen red and NIR wavelength there are different absorption coefficients, one could utilize Beer-Lambert law to calculate the concentrations of oxygen present in blood. A formula derived from this method is the "ratio of ratios" which compares the AC and DC components of detected red and NIR light to calculate ratio parameter for which SpO2 values of a subject can be calculated (1). The AC component arises from the volumetric blood flow causing an temporally varying extended optical path length as the microvascular blood vessels are dilated and contracted. The DC components are the result of a minimum path length when the blood vessels are at their most contracted, giving way to a baseline level of attenuation. The inclusion of the green wavelength was done due to it's ability to provide a resilience to a common artifact present in PPG technology, the motion artifact. Motion artifacts are the results of the sensor modules shifting it's position relative to the area of collection, resulting in noise that manifests as drastic DC baseline wandering, as well as added peaks in the signal that complicate the biometric extraction process. Green wavelengths have been shown to be have resilience to these artifacts in various products, and the origin of this resilience has recently been proposed to be the result of the green PPG signal originating from the muscle contracts around the blood vessels, rather than the blood vessels themselves (2). The downside to utilizing green wavelengths is that they experience much higher attenuation, therefore higher optical power is required.

Given that for reflection mode the collected light is diffused light, a cross polarization configuration was implemented to reduce the collection of specularly reflected light

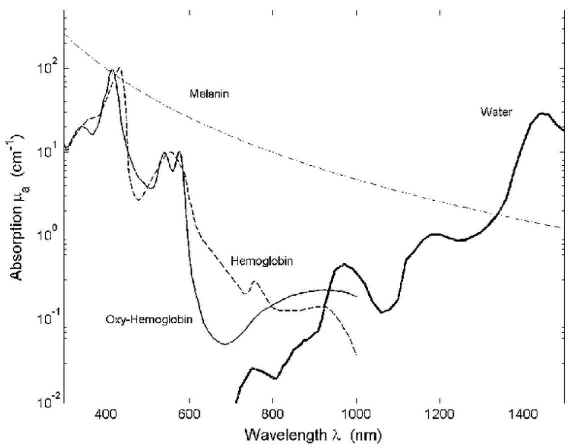

Fig. 2. Absorption Spectra of chromophores typically found in skin. Retrieved from (3)

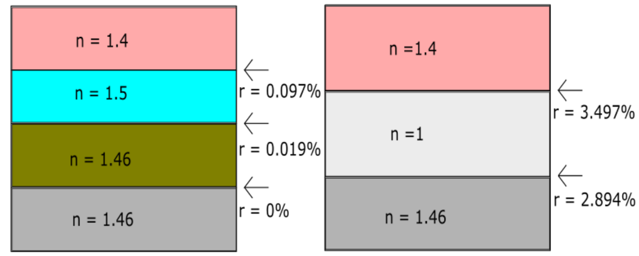

Fig. 3. Fresnel Losses from sensor

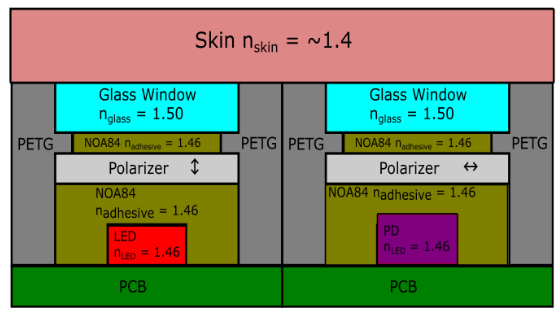

Fig. 4. Layout of sensor structure and components

that might be incident on the photo-diode contributing to an unnecessary DC offset. The polarizer used for this implementation was Thorlab's LPVISE2X2 linear polarizers which offer a 38% transmission for unpolarized light, and a cross polarization transmission of less than 1%. This implementation comes with the caveat of lower optical power emitted from the LED, therefore the forward current applied must be increased in order to compensate for the losses. In addition to the increases launching losses, the polarizer is not designed to operate for NIR wavelengths, therefore the SpO2 measurements must be calibrated in order to compensate for the reduced offset of values for the red wavelength.

To optimize light launching into the skin and subsequent collection, refractive index matching of the sensor to the skin was considered and implemented in order to reduce Fresnel reflections at sensor-skin interface. Fresnel losses can be modelled by the change in refractive index from one medium to another given the following equation.

$$r = \frac{(n_1 - n_2)^2}{(n_1 + n_2)^2}$$

where n_1 is the initial medium of propagation and n_2 is the final medium of propagation.

We utilized clear optical cast plastic windows from Edmund Optics as the skin contact structure to provide an index matching function as well as mechanical support. The chosen window offers a refractive index of around n_{window} = 1.501 and a transmission of over 90% for the wavelengths of operation. To adhere the optical windows to our sensor's components and structure barrier structures without introducing additional losses, Norland optical adhesive 84 was used. Norland optical adhesive 84 provides a refractive index of around $n_{adhesive}$ = 1.46 with a transmission of over 95% at the wavelength of operation. The refractive index of the adhesive also matches that of the silicone encasing of the LEDs and Photo-diode, $n_{LED/Photo-diode}$ = 1.46. Figure shows the layout of the medium interfaces for the device and skin. Given the stated refractive indices, the total losses attributed to Fresnel losses 0.12%, which is a notable improvement compared to the non-index matched arrangement, which provides a loss of 6.39%.

Opto-mechanical considerations were implemented to reduce DC noise resulting from optical cross talk of the emitter-to-detector without interacting with skin tissue. This was accomplished by introducing a barrier between the LED and photo-diode via the hand frame housing. The material used for the frame and the overall device enclosure was a PETG filament which provides less than 1 percent transmission of light at the operating wavelengths. The thickness of the barrier used was 1.5 mm (4). The arrangement of the optical components with opto-mechanical barriers can be seen in Figure 3.

The optical thermal sensor used on the device is MLX90632 by Melexis which offers medical grade accuracy of 0.01 degree Celsius. For the implementation of the MLX90632, an unobstructed viewing angle of 40° is required to achieve adequate sensitivity. This was implemented by introducing a spacing 1.68 mm between the thermal sensor and the hand frame.

AFE.

For our device, we have implemented six LEDs and two photodiodes divided into two separate sensing modules each composed of three LEDs of differing wavelengths and a photodiode. With each sensing unit configured in a parallel arrangement we will be able to achieve multi-directional light sensing. The implementation of two photo diodes receiving input from a total of 6 LEDs in a round robin fashion is beyond the capabilities of a low-profile microcontroller on its own. Compact Micro-controllers on their own do not feature very high-resolution analog to digital conversion, and are limited in their ability to sample from two different photodiodes accurately. To facilitate the integration of all the LEDs and photodiodes we employ an Analog Front End integrated circuit to manage the sensing units on their own and deliver the conditioned digital signal to the Microcontroller.

The Analog Front End used in this device is the AFE4404, shown in figure x below, from Texas Instruments, and it is the driver for three LEDs of different wavelengths and receive the corresponding current signal from a photodiode. It is programmable with an LED current of up to 50mA by default, but the current sinking limit can increase to 100mA per LED by configuring the registers, at the cost of accuracy in the exact current going through the LED. When the photo-diode captures light and translates it into a current signal, the current signal is a composite of three signals: an AC component that correlates to changes in blood volume, a DC component that is generated by the light reflected from the body's time invariant systems, and ambient light.

The current signal from the photo-diode is then converted to a voltage level through the means of a transimpedance amplifier. The signal is then conditioned through the use of a digital-to-analog converter which attenuates the signal from the transimpedance amplifier according to the level of ambient light, with the aim of amplifying a voltage signal with less interference. Once the voltage has been amplified, capacitors store the LED reflected light value and ambient light values separately, where they are then sequentially converted to a digital signal.

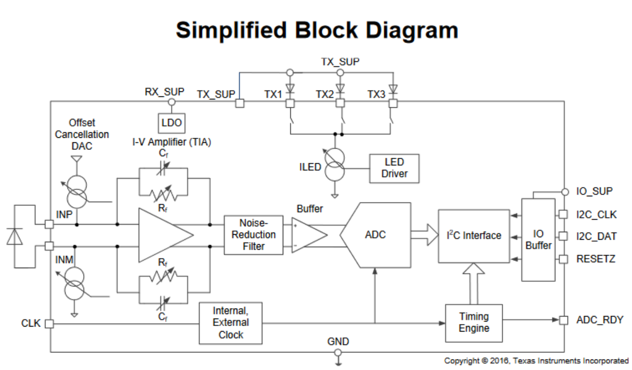

Fig. 5. AFE4404, diagram courtesy of TI

The resulting output from the AFE4404 is a composite light signal, ambient light, and the isolated voltage signal which is composite signal minus the ambient light. The device has an Analog to Digital Converter with a resolution of 22 bits and works in a range of -1 to 1 Volt. For accurate SPO2 measurements, the device must operate at a dc level of approximately 50 percent the positive voltage threshold. The output from the ADC is then captured in four 24-bit registers which store the resulting LED1, LED2, LED3, and Ambient voltage values in two's complement format. These registers are then read through I2C by our microcontroller.

Micro-controller.

Our device will be using a microcontroller that hosts an ARM Cortex-M4 32-bit processor, which boasts a 64 MHz clock speed, 1MB flash memory, and 256kB of RAM. This processor is integrated into the nRF52840 microcontroller which itself is on a NINAB306 module. The NINA module combines the nRF52840 microcontroller with an antenna to enable Bluetooth connectivity, which will be the main method of delivering our biometric data. Our microcontroller is tasked with receiving the PPG data over I2C from the AFE4404, checking that the data is valid, and then transmitting it over Bluetooth where it will be interpreted by our mobile application.

Power.

The battery we have chosen is the Lithium-Ion Polymer 500 mAh from PKCELL. It has a nominal voltage of 3.7V and an outstanding energy density providing 500mAh in a 29mm x 36mm x 4.75mm form factor. The Wearable Optical Biometric Tracking System will never exceed one Amp in current draw and is unlikely to ever get close. In terms of peak current draw, each AFE and LED combination consumes approximately 1 to 3 mA, the thermal sensor consumes 2mA, and the NINA can draw up to 40mA. A conservative estimate of current consumption places it at under 55 mA at peak load, suggesting the device is likely to have approximately 10 hours battery life if heavily used with the LEDs pulsing at high currents.

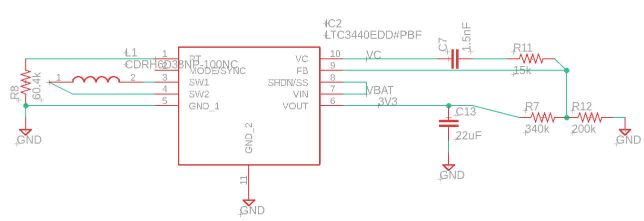

Fig. 6. LTC3340 Buck-Boost Converter schematic

The entire system will be powered by the lone battery, which will supply its nominal voltage to the SFH7016 LED units, and have its voltage converted to 3.3V to supply our microcontroller and provide the 3.3V logic level necessary for the system to function. The voltage conversion will be handled by a buck-boost DC/DC converter, the LTC3340, setup in a configuration that outputs 3.3V as seen above in figure 6.

IV. Signal Processing

Our device's primary goal as a biometric sensor is to accurately and consistently map a vector of input parameters to a set of predictions. This is pretty straight forward however it's worth understanding the reasoning of why we think this is possible. That is why this chapter is dedicated to detailing the logic driven processing algorithms our device will be using.

Peak Detection. One of the most important features we will be looking for is the signal's peaks. By taking a look at the sample PPG signal you will see there are two local maximums per heart rate period. We specify that we want to locate the highest valued maximum to maintain consistency

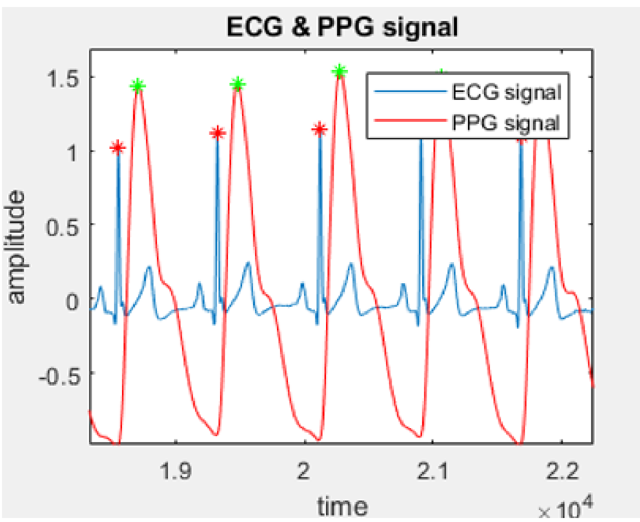

Fig. 7. Peak detection algorithm applied to sample PPG data.

Average Heart Rate. By detecting multiple instantaneous heart rates in series we are able to make an estimation for the average heart rate. This is important to note since not only will this parameter be reported to the user, but will also be used to estimate heart rate variation.

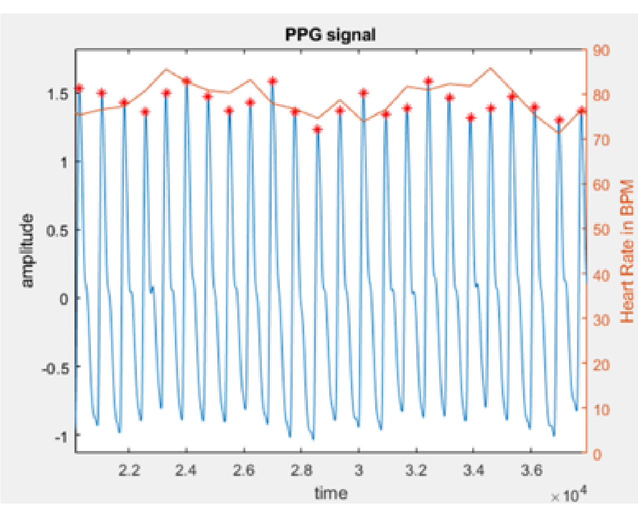

Fig. 8. Average Heart Rate over time using example PPG data.

Heart Rate Variation. Though there are many different ways to calculate heart rate variation, the most suitable for our application would be using Root Mean Square of Successive differences. RMSSD simply requires the interval between each of the peaks we capture. This is appropriate since we already have this data to compute the average Heart Rate. Below is the formula for RMSSD as well as a plot across time using sample data.

$$HRV = \sqrt{\frac{((Tn - Tn - 1)^2 + (Tn - 1 - Tn - 2)^2 + ...)}{N}} \tag{1}$$

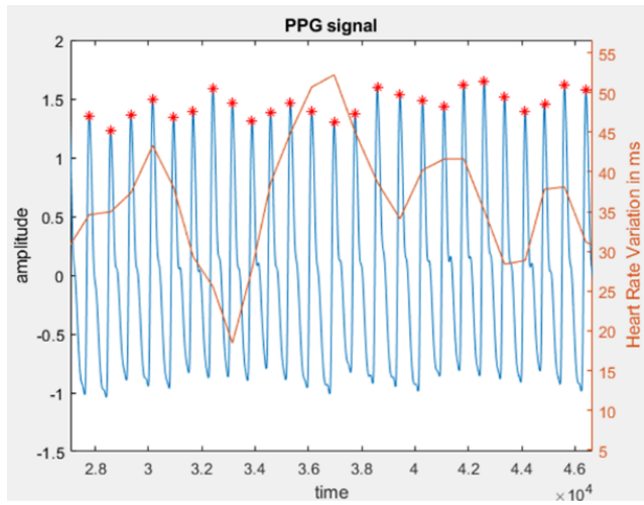

Fig. 9. Heart Rate Variation plotted across time.

AC and DC Component

The DC component is the average value across the period of the signal. The dominant AC component on the other hand is just the highest magnitude non zero frequency component. By taking advantage of the fast Fourier transform we can quickly identify both of these parameters. One thing to note is that the FFT can give you a better result the longer the signal you pass it. This also means that it will consume more computing power. Therefore it is unlikely this will be able to be calculated real-time. The discrete Fourier transform is defined below.

$$c_k = \frac{1}{N} \sum_{j=0}^{N-1} f_j e^{-ijk2\pi/N}, \quad k = 0, \ldots, N-1. \tag{2}$$

The Fourier transform will map a time domain signal to the frequency domain. The image below is a plot of the frequency components of sample IR and Red PPG signals.

Blood Oxygenation. The proper technique for predicting blood oxygenation is a debated topic. As of now we will be using the simplest approach, which is a ratio of ratios. AC_{red} and AC_{ir} refers to the dominant(global maximum outside of DC component) non DC component of the signal.

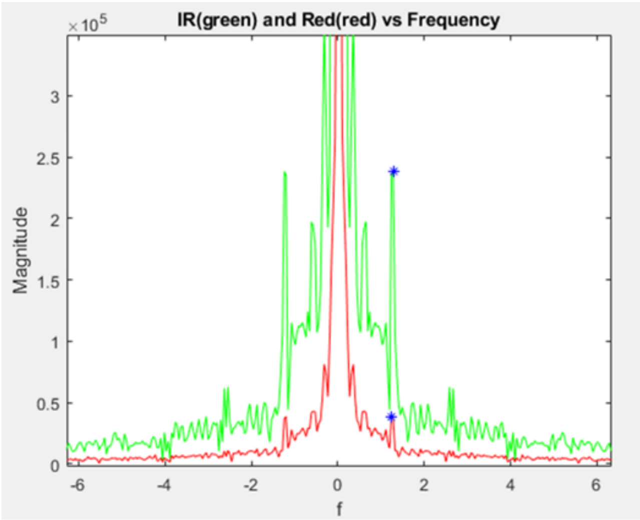

Fig. 10. Plot of Fourier Transform operating on a Red and IR PPG signal.

$$R = \frac{AC_{red}/DC_{red}}{AC_{ir}/DC_{ir}} \quad \textbf{(3)}$$

$$SpO_2 = C - \beta(R), \text{where C and } \beta \text{ are constants} \quad \textbf{(4)}$$

Though this technique is very simple it requires that the device be calibrated to find the constants. This implies that we can create a system of equations using a trustworthy device.

Signal Conditioning

Given that the human body operates at a relatively low frequency, most of our interest lies within the low frequency band(1Hz-5Hz). By sampling at a high frequency and applying a low pass filter we can remove any of the higher frequency components that would cause our heart rate prediction algorithm to fail.

Though a basic moving average would work fine here as a low pass filter, we decided to take this as an opportunity to be a bit clever. By using a Cascaded integrator-comb filter, we were able to build the exact same low pass filter, which not only uses a fraction of the hardware to implement, but also reduces our sample rate without the loss of information in the pass band. This is commonly called an integrate and dump filter due to its use of a single register and single adder.

According to the described structure the output of the accumulator appears to be a saw tooth wave. By sampling at each peak in the saw tooth wave we effectively create a digital low pass filter. With the following transfer function.

$$H(f) = \frac{sin(\frac{\pi}{2} * \frac{f}{f_s/d_s})}{f} \quad \textbf{(5)}$$

d_s: down sampling ratio
f_s: Sample Rate

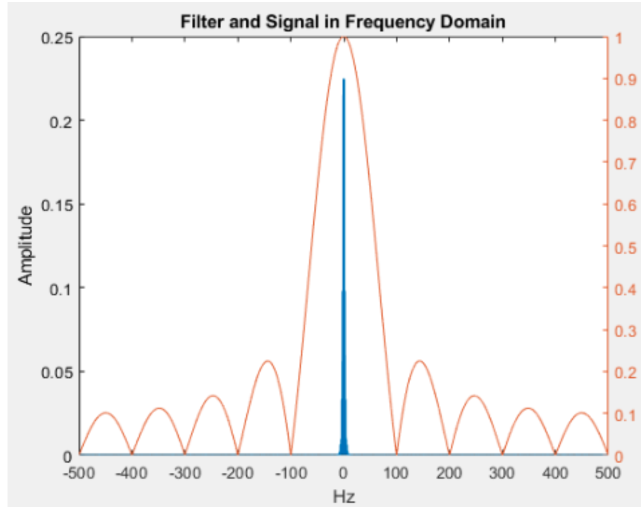

Fig. 11. Filter Transfer function and Example PPG signal frequency domain plot.

IV. Application Interface

Convenience is paramount when attempting to develop a system to be used as a fitness tracker. Someone utilizing this device may not be able to hook up a computer where ever they are to mark down information during exercise or research. Further a small screen on a wearable device may not provide sufficient viewing space to review progress or historical data for comparison. When considering how to interface with a device like the one developed here it has to be considered that computational power or storage space may be required and it must not be inconvenient for the user.

This desire for convenience of the intended application has informed nearly every aspect of the application interface design. In the interest of not wanting users to need much additional hardware and to allow a measure of freedom this design will leverage a mobile device.

On this same vein, although connectivity is getting better, not all locations afford network or internet connectivity. Implementing a local storage solution allows users to interact with the system even if they are in an area with spotty coverage or lacking in internet or mobile connectivity.

Mobile Phone Application. The compliment to this system is going to be a mobile application hosted on an Android device. A mobile computer with built in security features as well as viewing frame larger than could be expected to put on a wearable device that people are already carrying around is an obvious choice. Although either Android or IPhone could have been the launching point, the decision was made to start on Android for this prototype due to many factors including initial cost, availability of development tools, and market share. The sensor device will leverage the mobile processor in the mobile platform in order to process sensor data. This will allow the device to be lighter and smaller since it will leverage the mobile CPU with the signal processing algorithms the storage capacity and database support to capture and store new readings. This also affords users a familiar interface with built in security to review historical data. The signal processing algorithms will be incorporated as part of a custom application built that will connect over Bluetooth Low Energy to our device which is capturing the data.

Connectivity. In the interest of compatibility the application is designed to communicate over the Bluetooth Low Energy standard. This is a protocol supported by many mobile platforms and is an improvement over the original Bluetooth standards due to its nature of frugal energy consumption. This means less battery changes or charging for users on their sensing device or mobile phone. Using the Bluetooth Low Energy protocol data will be provided by the device as Characteristics presented by a custom service provided by the custom sensing device. From there the data will be consumed by the application. The application at this point utilizes the Signal Processing algorithms described above in order to extract the biometric data which is stored.

Application Flow. The application is designed to support multiple users access to the system utilizing a Login and passcode combination. This login will allow the system to be segregated by user account, either separating data for specific events or allowing multiple users of the application. Once logged in the user will have the option of either reviewing stored data or capturing new data to analyze.

If the user decides to capture new data it interfaces with the custom appliance to receive this information to inform the biometric measurements. This is where the bulk of the signal processing comes in, calculating biometrics from the shaped sensor data. If instead they decide to review historical data it will lead to another area of the mobile application where data can be retrieved.

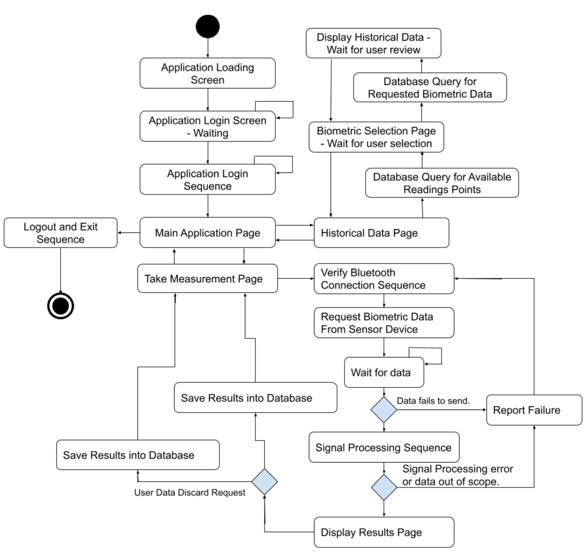

Fig. 12. State diagram of the logical flow of the mobile application.

Data Storage. Once the data is processed into the biometrics it is then captured by the local instance of our database system. The database system employed here is SQLite. The structure of SQLite makes provides a local interface to store the data without the need for any external devices or a network connection. It also lends to the security of the data, as it can only be accessed by one with access to the mobile device.

If a user decides to view historical data instead, they will be presented with a list of previously captured events and provided the stored boimetrics which relate to those events. The database structured into 6 tables holding own unique data. Being a SQL structure, the data must be uniform to be easily accessed through structured queries.

Users		1
id*	username	password
Integer	String	String
Primary key, Inc	unique	

Sensor_Function		2
id*	blueToothClassCode	Description
Integer	Integer	String
Primary key, Inc	Unique	

Biometric_IDs		3
id*	Units	Description
Integer	String	String
Primary key, Inc		

Recording_Session			4
id*	UserID	Start_time	End_time
Integer	Integer	String	String
Primary key, Inc	Foreign key (1)		

Sensor_Readings_Raw				5
id*	UserID	Sensor_ID	TimeStamp	Reading
Integer	Integer	Integer	Date_time	Real
Primary key, Inc	Foreign key (1)	Foreign key (2)		

Derived_Biometrics				6
id*	UserID	Biometric_ID	SessionID	Value
Integer	Integer	Integer	Integer	Real
Primary key	Foreign Key (1)	Foreign Key (3)	Foreign Key (4)	

Fig. 13. Table Structure of the storage database implemented in SQLite

Conclusions

The theory discussed in this conference paper was the basis of our final device, but a fraction of the effort. The sensitive nature of this application made the implementation a much more challenging endeavor. If more time was allocated to this project, more time would have been spent researching ideal locations to measure on the wrist as well as developing models to predict ideal transcendence amplifier feedback settings given component placement and housing parameters.

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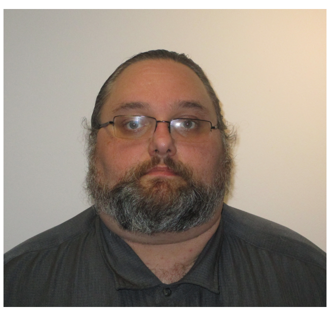
Biography

Scott Perryman is currently a senior at University of Central Florida and will receive a Bachelor's of Science in both Electrical and Computer Engineering in August of 2022. His primary interests lie in Signal Processing and control systems. Upon graduation, Scott plans to join The Mitre Corporation.

Cody Showers is a Computer Engineering Senior at the University of Central Florida. Currently he is working as an Information Technology professional at Siemens Energy at their US home office location. Upon completion of his studies he intends to pursue a career in the Computer Engineering profession.



Christopher Bormey is a Senior graduating with a bachelor's in electrical engineering. He will be searching for an Electrical Engineering position to work at for the near future before going back to pursue higher-level education.

Tyrone Morales is currently a senior of the Photonics Science and Engineering Program at the University of Central Florida. He currently works with Nanophotonics Material (NPM) research group at CREOL and plans to pursue a graduate degree in the near future.