# Skin Color-Corrected Pulse Oximetry

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**Abstract**

Pulse oximeters have exhibited inaccuracies in measuring blood oxygen levels for individuals with a darker skin tones, potentially contributing to healthcare disparities based on race. The research aims to enhance the standard Pulse Oximeter to mitigate this discrepancy, thereby improving the well-being and medical care for individuals with diverse skin tones. This project investigates various methods of Pulse Oximeter utilizing a multi-wavelength spectrometer to measure the patients skin tone. Additionally, we are investigating how to use more than the standard two wavelengths (red and infrared light) to measure blood oxygen levels accurately.

## 1 Introduction

In response to an alarming discrepancy in existing Pulse Oximeter technologies, particularly the accuracy of blood oxygen level measurements in individuals with dark skin tones, this projects goal is to develop and advance the existing model. The focus is on creating a Pulse Oximeter capable of accurately and reliably assessing blood oxygen levels in patients with darker skin tones. Recognizing the significant consequences of inaccurate readings, especially within the communities of color where fatal outcomes an result from such errors, this project is driven to address and rectify this critical issue. This issue has been in our medical community since reading blood oxygen levels was done through skin.

The science of noninvasive oximetry started a about a century ago. The most accurate way to measure blood oxygen levels is through arterial blood samples taken from a patient. However, this process is time consuming and can put unneeded stress on patients. Thus a noninvasive method of reading blood oxygen levels was sought after. The first oximeters came into wide use during WW2, utilizing red and near-infrared light (exemplified in figure 1). The next major

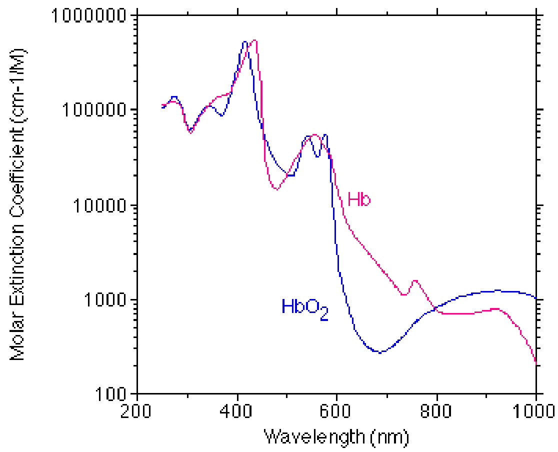


Figure 1: The Change in the Extinction Coefficients of Oxygenated and Reduced Hemoglobin for Different Wavelengths. Graph sourced from reference [8]

development in oximetry would come in the early 1980's with the invention of the Pulse Oximeter, which utilized a patients pulse to distinguish between arterial blood and other parts of the body. Medical professionals assumed that since this method is supposed to be only measuring the substances in arterial blood, it effectively filtered substances like melanin. This proved to be untrue for numerous reasons, as will be discussed later. Yet, a lost "Gold Standard" glitters from the past, indicating what could have been. Developed at the same time as Pulse Oximeter, an eight wavelength oximeter designed by HP seemed to solve the skin color bias of oximeter's past. According to an article written in the October 1976 edition of the Hewlett-Packard Journal[5], their new oximeter had similar efficacy for Black subjects as for non-Black subjects. Developing a method to combine Pulse Oximeter with HP's multi-wavelength method thus became the focus of this project.

### 1.1 The Beer-Lambert Law

The Beer-Lambert law is the basis for all noninvasive oximetry. It describes the relationship between the optical attenuation of light through a material and the absorption of that light by multiple species of substances. In the simplest case, when there is a single species of substance of uniform concentration, the Beer-Lambert law can be written as:

$$\log \frac{I_0}{I} = -\log T = A = l\epsilon c, \tag{1}$$

where $I$ is the attenuated intensity, $I_0$ is the initial intensity, $T$ is the transmission, $A$ is the absorption, $l$ is the optical path length, $\epsilon$ is the absorptivity or extinction coefficient for the species of substance, and $c$ is the concentration of the species of substance. This law also states that for a material with multiple species of substances, the absorption is linear. Thus the more general formulation of the Beer-Lambert Law for wavelength $k$ and $m$ species of substances is:

$$-\log T_k = A_k = \sum_{i=1}^{m} A_{k,i} = \sum_{i=1}^{m} \epsilon_{k,i} \int_0^l c_i(z)\mathrm{d}z. \tag{2}$$

However, for the purposes of oximetry, we assume that the material is non-scattering (or at the very least minimal scattering), the concentrations are uniform, and the optical path length is the depth $d$ of the material for all wavelengths. These assumptions simplify the equation considerably:

$$A_k = d\sum_{i=1}^{m} \epsilon_{k,i} c_i. \tag{3}$$

For most Pulse Oximeters, only two wavelengths, red and near-infrared, are used to measure blood oxygen levels. The Beer-Lambert law equations for red and infrared with only oxygenated hemoglobin, represented by o, and reduced, represented by r, (equations 4a and 4b).

$$A_R = d(\epsilon_{R,o}c_o + \epsilon_{R,r}c_r) \tag{4a}$$
$$A_{IR} = d(\epsilon_{IR,o}c_o + \epsilon_{IR,r}c_r) \tag{4b}$$
$$\beta = A_R/A_{IR} \tag{4c}$$

$$S_i = \frac{c_i}{c_o + c_r} \tag{5a}$$
$$S_r = 1 - S_o \tag{5b}$$

$$S_o = \frac{\epsilon_{R,r} - \beta\epsilon_{IR,r}}{(\epsilon_{R,r} - \epsilon_{R,o}) - \beta(\epsilon_{IR,r} - \epsilon_{IR,o})} \tag{6}$$

By using the ratio of attenuation equations 4 and the hemoglobin saturation equations 5, a simple algorithm, equation 6, using a single ratio can be found. [9]

## 1.2 Putting the "Pulse" in Pulse Oximetry

The Pulse Oximeter method was developed in the early 1980's. It utilizes the systole beat of the heart to attempt to measure only substances in arterial blood. The systole is when the heart contracts, pushing blood through out the body. This causes more blood to be pushed through extremities like fingers. This causes the transmission to precipitously drop from its maximum to its minimum. It is assumed that this change in transmission is caused only by arterial blood. for each wavelength, two measurements are taken to find two

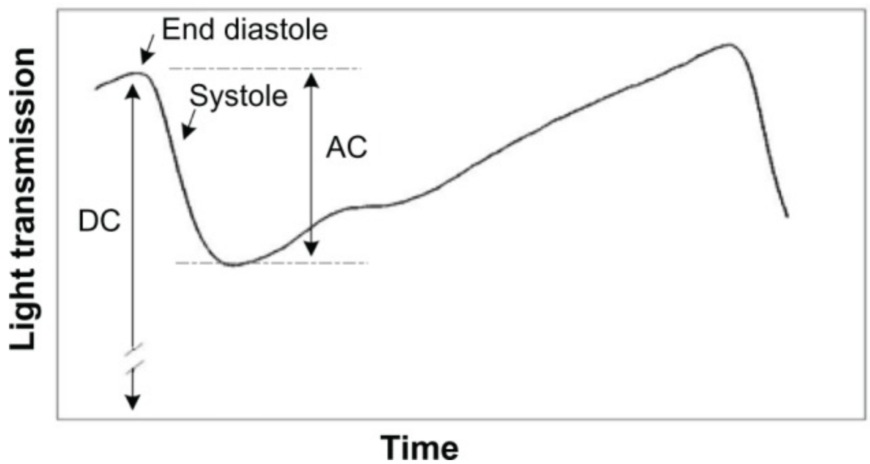


Figure 2: Example of the transmission signal for a single wavelength over the course of a single heartbeat. Sourced from reference [6]

components:AC and DC (as shown in figure 2. The DC component is the total absorption of all the substance. The AC component is the amplitude of the wave caused by the pulse. By using a ratio-of-ratios (equation 7a) instead of the simple ratio of absorption. Because this ratio-of ratios is substituting for the simple ratio, the explicit coefficients cannot be used. Instead, they must be found empirically for each pulse oximeter model (equation 7b).

$$R = A_{AC,R}/A_{DC,R}/A_{AC,IR}/A_{DC,IR} \tag{7a}$$

$$S_o = \frac{k_1 - Rk_2}{k_3 - Rk_4} \tag{7b}$$

### 1.3 A Lost 'Golden Standard': Making a Multi-Ratio Pulse Oximeter

$$S_o = \frac{a_0 - \sum_{n=1}^{8} a_n A_n}{b_0 - \sum_{n=1}^{8} b_n A_n} \tag{8a}$$

$$A_n = -\log T_n \tag{8b}$$

$$0 = \sum_{n=1}^{8} a_n = \sum_{n=1}^{8} b_n \tag{8c}$$

HP used Kramer's method of solving systems of equations to find a formula for blood oxygen saturation using 8 wavelengths of light as shown in equation 8. The $a_0$ and $b_0$ are mathematical artifacts of setting the boundary conditions in equation 8c. Taking this coefficient out, the top and bottom con be divided by

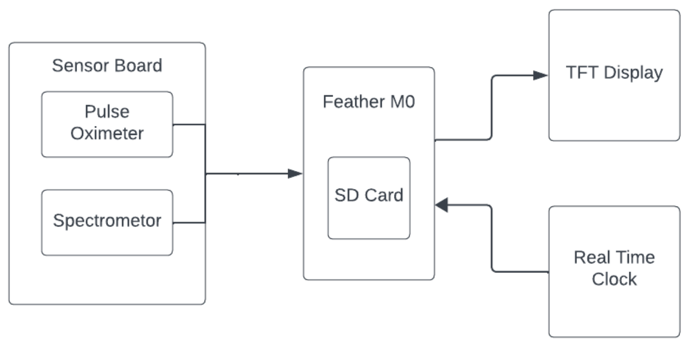


Figure 3: Simplified overview of the hardware layout. To the left is the sensor board and the right is the main board.

the absorption of a reference wavelength, creating equation 9. This equation is the basis for this projects pulse oximeter going forward.

$$S_o = \frac{a_0 - \sum_{n=1}^{N-1} a_n R_n}{b_0 - \sum_{n=1}^{N-1} b_n R_n} \tag{9a}$$

$$R_n = A_{AC,n}/A_{DC,n}/A_{AC,IR}/A_{DC,IR} \tag{9b}$$

# 2 Methods

## 2.1 Overview

We created two printed circuit boards connected together using ribbon cables: one for the sensors and one to display our results. The display board is comprised of an Adafruit Feather M0 Adalogger along with an Adafruit wide angle TFT LCD display withe a microSD connector, an Adafruit DS3231 Precsion RTC sensor, and a ribbon cable connection. The sensor board contains a Adafruit AS7341 11 Channel Light/Color Sensor, Spark-Fun Pulse Oximeter and Heart Rate Monitor (MAX30101 & MAX32664), and a ribbon connection to attach to the display board. These two boards work together to measure, process, present data with precision and accuracy.

## 2.2 Display Board Devices

### 2.2.1 Adafruit Feather M0 Adalogger

The Adafruit Feather is the microcontroller responsible for data processing and management [4]. This device also has features a built-in SD card slot which

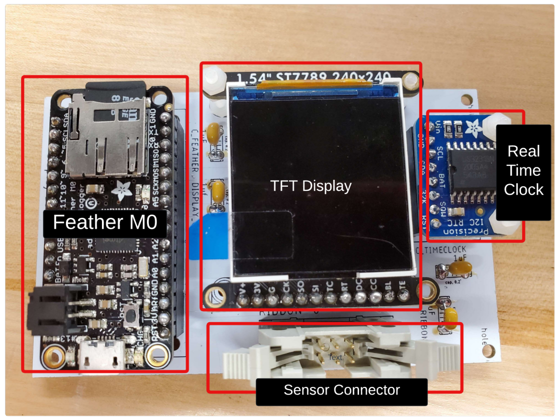


Figure 4: This figure shows the main display PCB board. It features the TFT LCD display, the Feather M0 Adalogger, the RTC sensor, and the ribbon cable connectors indicated by the red boxes.

allows for seamless data collection. In this project, we need high processing power for taking reliable and accurate measurements.

#### 2.2.2 Adafruit Wide Angle TFT LCD Display

This devices serves as a visual interface, providing a comprehensive and clear color presentation of our results. The integration of a microSD connector further extends the functionality, enabling additional data storage [1]. We want to include this display to see our blood oxygen levels and heart rate in real time displayed on the screen.

#### 2.2.3 Adafruit DS3231 Precision RTC Sensor

Including a Real-Time Clock (RTC) on this boards allows for precise timekeeping which helps to have an accurate understanding of when each measurement was taken. Our device is time dependent since the signal we want to measure is changing over time [3]. The oxygenation of a patients blood is analog therefore we need to keep timestamp track. We need our measurements to be at millisecond accuracy since a patients heart usually falls in the range of 60-100 BMP. This means we expect approximately 1 beat per second. With that information, this is why we need millisecond accuracy to track in real time the heart rate. Also, if patients have heart defects or other heart related conditions, we need to be able to track those symptoms with the Pulse Oximeter with quick measurements. The Feather M0 Adalogger does have a built in internal clock with millisecond accuracy. This internal clock is not calibrated with a real time source, therefore it only has a certain degree of accuracy. With the RTC used in conjunction with the Feather Adalogger, we can timestamp track with millisecond accuracy.

## 2.3 Sensor Board Devices

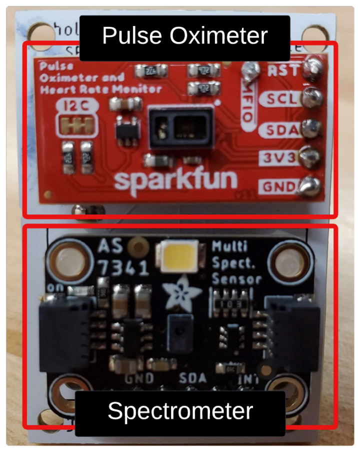


Figure 5: This figure shows the sensor board. It features the SparkFun Pulse Oximeter and Heart Rate Monitor and the Adafruit AS7341 11-Channel Spectrometer.

### 2.3.1 Adafruit AS7341 11 Channel Light/Color Sensor

The AS7341 Light Sensor serves as a crucial part of the project as it is designed to capture and interpret a broad spectrum of light and color data. With the 11 channels, this sensor enables detailed analysis of the environmental conditions, ambient light measurements, and spectral analysis [2]. Its purpose within this device is the actual measurement of a patients skin color. This is where our device differs from the average Pulse Oximeter. The AS7341 gives access to 7 more wavelengths (445nm-680nm) than a normal Pulse Oximeter, these additional terms can be taken into account when doing the oxygen saturation calculation to more accurately account for a patients skin color.

### 2.3.2 Spark-Fun Pulse Oximeter and Heart Rate Monitor

The inclusion of the Spark-Fun Pulse Oximeter and Heart Rate Monitor is imperative to the projects main goal. The MAX30101 sensor allows for real-time measurement of both Pulse Oximeter and heart rate. The MAX32664 further pushes the processing power ensures quick and efficient analysis of many physiological parameters [7]. We will use the data from this sensor to compare to our spectrometer data.

## 2.4 PCB Board

Once deciding on the parts that were going to be used in this project, a PCB board was created in the software EAGLE. This software let's PCB designers to seamlessly connect schematic diagrams, component placement, and PCB routing. As mentioned before, two boards were created for this project. The display board is the larger of the two boards. The placement of each component was important as the accessibility to the microSD ports and battery ports needed to be open for easy use. See Figure 2 for the display PCB board.

For the sensor board, a compact footprint was important, given its role in measuring patients' physiological data through finger-based monitoring. See Figure 3 for sensor board. The design prioritized space efficiency without compromising the functionality of the embedded sensors. The two boards are connected with a ribbon cable so the measurements from the sensor board can be displayed and then recorded to the microSD on the display board.

## 2.5 3D-Printed Finger Case

In conventional Pulse Oximeters, a prevalent design employs an alligator clamp mechanism, typically affixed to the patient's index finger. In this project, the objective diverges from the traditional approach of reading transmitted light through the finger. Instead, our focus lies in precisely measuring the light reflected from the underside of the finger. Additionally, during the evaluation of the spectrometer and Spark-Fun Pulse Oximeter sensors, it's imperative to note that variations in applied pressure can significantly impact data accuracy.

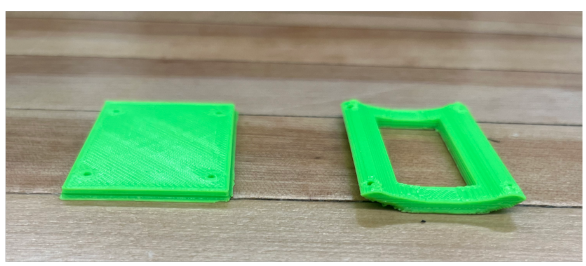

Figure 6: This figure shows the 3D-printed finger holding case. The part on the right is the device to hold the finger steady. The green board on the left is for stabilizing the system once assembled. There will be 4 screws going through the corners of each part pictured. The sensor board will be sandwiched between the two parts. Nuts will be places under each component once assembled for easy height adjustment of the finger case over the sensor board for accurate readings.

Optimal performance is contingent on maintaining a delicate balance in pressure, as excessive or insufficient force on the sensors may compromise the accuracy of data acquisition. In order to conduct precise testing of the updated version of the Pulse Oximeter, a finger holding case needed to be fabricated. This case was designed with to guarantee a consistent and invariant distance between the finger and the sensors across diverse patient scenarios. This standardization is crucial for ensuring accurate and reproducible measurements during the testing phase. See Figure 4 for the 3D-printed finger holding case.

At the time of constructing the finger case, the correct distance from finger to sensor was unknown so the case was constructed with four holes on the corners. These holes line up with the sensor PCB so once 2 mm screws are threaded through the holes, the distance between the two can be adjusted with nuts. Once the optimal distance is decided, a new case can be printed with any other feedback gathered from the initial design.

## 2.6 Software

### 2.6.1 Software Overview

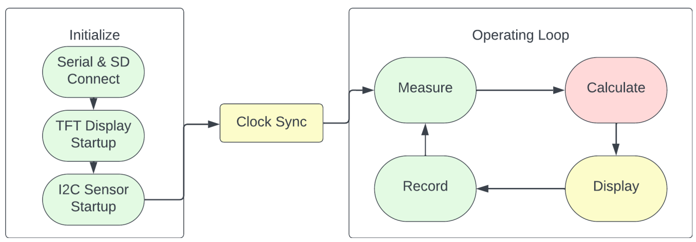


Figure 7: General organization of the software setup.

Figure 7 describes the organization and workflow of the software. It begins with an initialization phase that runs through the setup of the device. This is followed by a clock sync between the DS3231 real time clock and the Feather M0 internal timer. The DS3231 yields second accuracy and the Feather M0 yields millisecond accuracy, with the two combined the data has timestamps down to the millisecond which can be used to take multiple data points a second. With the setup complete, the operating loop will run indefinitely.

| **Library** | **Author** |
|---|---|
| DS3231.h | Eric Ayars |
| SPI.h | Aurduino LLC. |
| Arduino.h | Aurduino LLC. |
| Wire.h | Aurduino LLC. |
| SD.h | SparkFun Electronics |
| SparkFun Bio Sensor Hub Library.h | SparkFun Electronics |
| Adafruit AS7341.h | Adafruit |
| Adafruit GFX.h | Adafruit |
| Adafruit ST7789.h | Adafruit |

Table 1: Included libraries imported within the initialization of the code.

The device specific functionalities are written inside of dedicated libraries written by the manufacturer. This allows us to compose their functionalities easily. The specific libraries and which device they work with are shown in table 1.

### 2.6.2 Initialization of Code

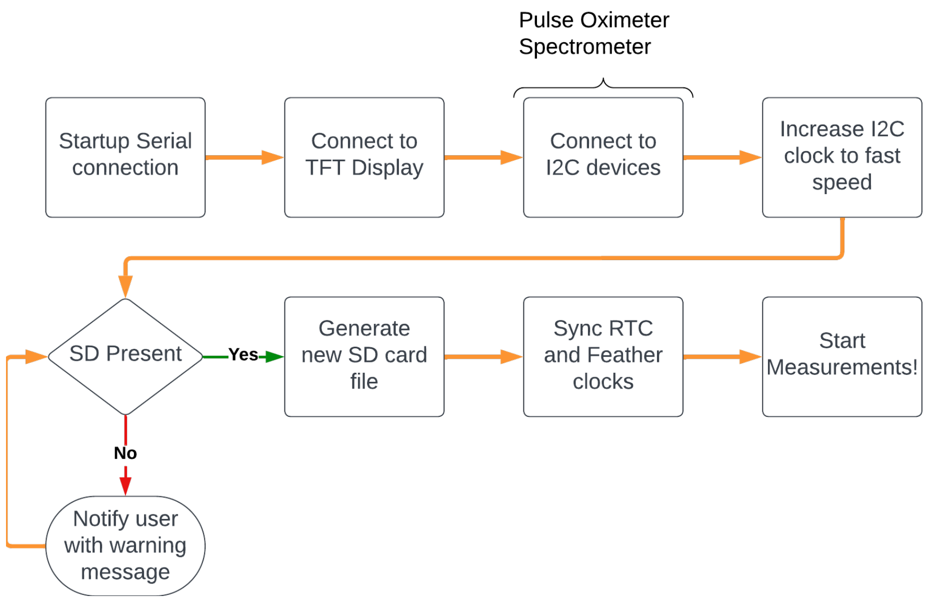


Figure 8: A diagram showing the initialization process of the device

The device initialization, as depicted by Figure 8, begins with connecting the device to a computer over a serial connection with a baud rate of 115200. Next the TFT Display is connected to using the SPI interface of the Feather M0. After this the I2C based devices, the Pulse Oximeter, the Spectrometer, and the RTC, are connected to and setup. In order to increase the rate at which measurements are taken the I2C speed is then increased from standard mode (100 kbits / second) to fast mode (400 kbits / second).

Once the communication rate is increased the SD card attached to the Feather M0 is detected. If it is not present then the user is notified on both the TFT Display and the serial connection. This continues until an SD card is detected. At which point a new data CSV file is created with columns for all I2C devices.

The final step in the initialization process is to sync the RTC with the internal clock of the Feather M0. This is done by waiting until the RTC has elapsed a single second. Once this happens the current value of the internal clock is recorded. This is used as an offset subtracted from the internal clock's value whenever it is recorded. This gives an approximate synchronization between the two clocks.

### 2.6.3 Operating Loop

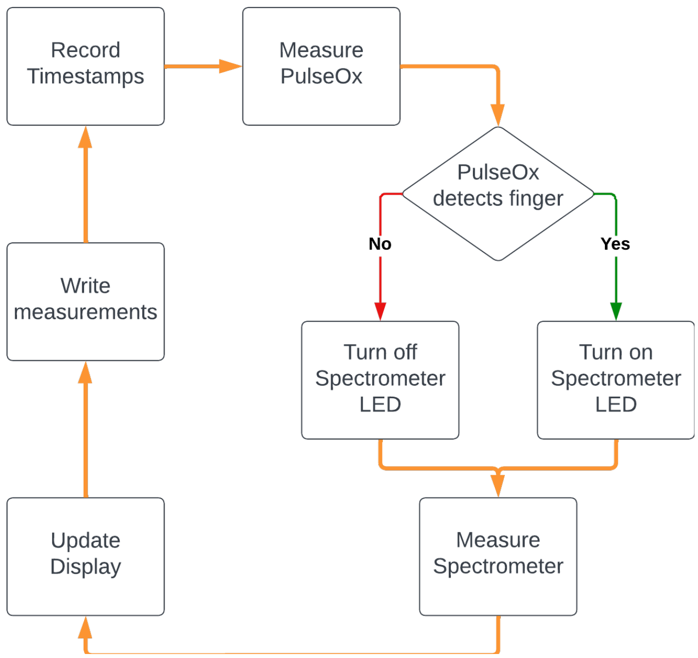


Figure 9: A block diagram of the operating loop of our device. The time stamps are intentionally taken as close to the measurements as possible to get accurate timing. The Spectrometer has an LED which is turned on or off depending on the data from the Pulse Oximeter.

The primary operating loop is a sequential but infinite loop that repeatedly takes data from the I2C devices as fast as possible while still updating the displays. It begins by getting timestamps from both the RTC and the Feather's internal timer. Then the Pulse Oximeter and Spectrometer are queried for data using their I2C interface. This data is currently only used to update the displays and then write the measurement to the CSV data file. At which point this loop now repeats.

## 2.7 Data Acquisition

In order to test the functionality of this device we need a standard experiment to acquire results from. The most basic functionality of our device is to measure the changes in the optical properties of the skin due to blood flow. This can be accomplished by taking measurements from both devices and waiting until the Pulse Oximeter successfully reads an oxygenation level. Once the Pulse Oximeter successfully measures we record the spectral counts from the Spectrometer and the measurements from the Pulse Oximeter for two minutes. For consistency, each participant used their right pointer finger and placed it on both modules with the tip of the finger on top of the Pulse Oximeter. Once the two minutes was up the CSV data file is copied onto a computer and analyzed using various python libraries.

# 3 Results

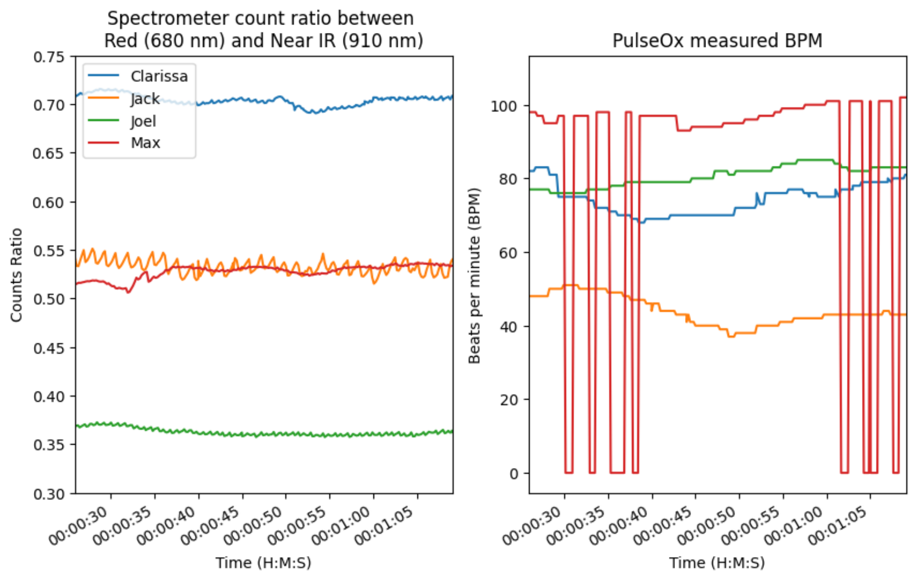


Figure 10: The left graph shows the ratio of two channels from the spectrometer for the two minute duration. The ridges of these lines indicate that the spectrometer is detecting the heartbeat of the subject. The right graph shows the measured BPM from the Pulse Oximeter module. The dropouts are due to the module's sensitivity to pressure

The counts ratio-time graph (left) shows the ratio of Red/IR light which is analogous to the ratio of $DC_{Red}$ / $DC_{IR}$ . The importance of this graph lie within the theory of calculating oxygen saturation. Oxygenated hemoglobin absorbs greater amounts of IR light and deoxygenated hemoglobin absorbs greater amounts of red light, this means the ratio is a part of the deoxygenated hemoglobin/ oxygenated hemoglobin ratio. All that's missing is the AC component of the hemoglobin ratio. That ratio is the basis of the calculation for oxygen saturation, the purpose of this device, so showing the counts and the devices ability to make this calculation is of the upmost importance.

Another important note is the visible heartbeat along the ratio line, this is most visible in the reading for Jack. This shows that even with the reduced ratio the device can still get basic readings. This shows that we can see lower frequency measurements, but if there were more specific heart deviations they would be too high frequency to see in the devices current state.

The reason behind Jack's heart rate spikes being more visible is due to the clearer peaks that show up in lower heart rate readings. Looking at the Beats Per Minute Graph (right) you can see Jack's heart rate tends to be lower than the others.

# 4 Discussion

## 4.1 Clock Sync

The clock sync is the product of the DS3231 not having millisecond accuracy, it records time down to the second in YEAR/MONTH/DAY/HR/MIN/SS format. Because this devices records multiple measurements per second, it requires the utilization of the Feather M0's millisecond timer to get the exact time of a measurement. Syncing the timers up so the Feather M0 is recording the milliseconds between each DS3231 second is the purpose of this point in the software. It is possible this process could yield a small uncertainty in the measurement accuracy of the timing because the time is being tracked with two sources as opposed to one all-encompassing timing module.

## 4.2 Data-taking speed

A primary outlet of discussion is the speed at which data is taken for this system. Because all the components rely on each other for each measurement, one component being slower than another can create a bottleneck for the whole process. Referring back to the operating loop diagram in figure 9, the loop is only as fast as each of the four steps.

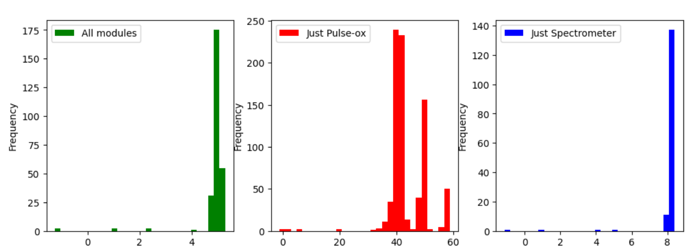


Figure 11: Measurement frequency graphs for each of the main modules of our device.

Figure 11 shows the measurement frequency for the entire device, the Pulse Oximeter module, and the spectrometer module. The Pulse Oximeter is the fastest, recording more than 40 measurements per second, but because the spectrometer only does 8 measurements per second, the loop is slowed from 40 to 8 measurements. When the calculation, display, and recording portions of the operating loop are added in, this slows even more to approximately 5 measurements per second.

In order to accurately recreate an analog signal the measurement frequency needs to be at least twice that of the maximum signal frequency by the Nyquist-Shannon sampling theorem. In figure 11 the vertical line indicates a lower bound of this sampling rate assuming that the heart beat itself is the highest frequency signal at 100 beats per minute. This frequency is approximately 3 samples per second which our device measures barely above. However, there are likely higher frequency components to blood flow due to inconsistency in the signal; such as irregularities in the heartbeat. Thereby our device can currently measure basic approximations of the oxygenation level but is not sensitive to higher frequency issues which would be necessary in hospital settings. Future iterations of this device should focus on increasing this measurement speed to become more sensitive to these high frequency signals.

### 4.3 Future Plans

The primary place for future improvements will be the full implementation of the the theory section. Currently, the device functions more as a proof of concept, having the ability to read skin color and pulse but without the necessary algorithm to alter the pulse readings based on skin color. The theory is in the late stages of development for the first usable algorithm that will be implemented over the next few weeks. Once it's ready, our groups main focus will be testing this algorithm.

Another improvement in development is our prototype version 2 PCB. This version 2 mainly consists of small improvements such as pin hole adjustments and fixing small errors the first board had. As we continue to use the two PCB boards together, we are keeping track of any additional changes that can improve efficiency.

We will also create another version of the finger holding case once we implement and test using the new theory. With the addition of the theory, we can then assess an improved finger case. We may choose to adopt the standard design of the finger clamp if we feel that produced the most accurate results. We may improve upon our current design where we lay the finger in the 3D printed case as seen in Figure 6.

As for software, we plan on correcting any time syncing issues between the RTC and the Feather Adalogger internal clock. The speed of the Spark-Fun Pulse Oximeter and the spectrometer data results needs to increased. We are implementing software functions to increase the speed of data collection which will increase our accuracy.

## 5 Conclusion

In this paper, we presented our current plan and developments on updating the commonly used medical device, the Pulse Oximeter. We have recognize the relevant inaccuracy of standard Pulse Oximeters as presented in the literature. Our plan includes exploring a wider range of wavelengths to observe the absorbance, reflectance, and transmittance of a finger to measure the blood oxygen levels. In addition to the wider wavelengths, which will be measured with the spectrometer, we also implemented the Spark-Fun Pulse Oximeter and Heart Rate Monitor to compare our results. Once we implement the equations from the theory section, we will able able to use make ratios of multi-wavelengths.

As for hardware, two PCBs were created to work together connected with a ribbon cable connector. One board displays our measurements and collects the data and the other board is taking the measurements. Minor adjustments are being made to improve the efficiency and layout of the PCB design. Additionally, we created a finger holding case to enhance precise and accurate Spark-Fun Pulse Oximeter results across all subjects. This case will be improved once we implement the theory.

## 6 Acknowledgements

We would like to thank the project staff: George Gollin, Yuk Tung Liu, and Shengzhu Yin. We would also like to thank Spark-Fun for providing the Pulse Oximeter device and the Carle Medical School for future collaborations in data testing on patients.

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