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Handheld Device for Remotely Measuring Brain Function

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ABSTRACT

Infants, and especially premature infants, are carefully monitored while in their incubators, and their brain health is of great concern. While it is currently the preferred procedure to attach a device to a patient's forehead for monitoring pulse rate and oxygen levels in the brain, the development of a remote handheld system would make it possible to spot check patients from a distance, without having to deal with the obtrusiveness of attaching probes. Additionally, past studies have shown that measuring changes in blood volume and oxygen levels in the forebrain can be used to study brain function. Current work suggests that affordable and safe handheld devices for contact systems can be built from inexpensive components such as LEDs and photodiodes to measure these parameters.

The goals of this project were to study the aspects of, analyze the feasibility of, and suggest a design for a handheld device capable of remotely sensing brain function by employing the basic building blocks of near-infrared technology. As part of the project, experiments have been designed to simulate the uses – such as finding the arterial pulse and tracking changes in blood volume and oxygen levels in the brain – of a handheld remote sensing device for measuring brain health and function. In this study, remote sensing has demonstrated promising results for use over small distances. This paper also includes suggestions for extending remote sensing over greater distances for use in more practical, real world situations.

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1. INTRODUCTION

Approximately nine percent of babies are born prematurely, and many need special care in a Neonatal Intensive Care Unit (NICU) [1]. Since they are not fully developed, premature babies frequently suffer from health problems and are carefully monitored in their incubators. Common health concerns include apnea, anemia, respiratory distress syndrome, and patent ductus arteriosus, of which pulse rate, blood volume, and oxygen levels can be very significant indicators [2]. Furthermore, several recent studies have suggested that premature babies are impacted negatively by noise, light, and activity in their NICUs. Thus, many hospitals take care to maintain a quiet environment, shield the babies from light, and handle them slowly and deliberately [1]. A handheld remote sensing device for measuring a baby's brain health parameters such as blood volume and oxygen levels would provide a more discrete and desirable method for assessing these health indicators..

In addition to monitoring the brain health of babies, such a device could also be used to evaluate other brain functions. Current devices use contact methods for monitoring brain function during tasks such as problem solving or lying. A probe placed on the subject's forehead measures changes in blood volume or oxygen levels in certain regions of the brain [3]. These changes correspond to brain activity in the region of the brain measured. If, for example, a certain region of the brain displays increased blood volume after a subject tells a lie, monitoring this region for changes in blood volume can determine whether the subject tells a lie again. Moreover, determining which parts of the brain are more active, as determined by changes in blood volume or oxygen levels during certain tasks, can map out overall brain function. Past tests include determining which areas of the forebrain show activity during lying or solving anagram puzzles. A handheld device for remotely measuring brain function would offer portability and convenience to both the user and the subject.

2. BACKGROUND

Near-infrared (NIR) light has been used in recent years to non-invasively measure optical properties of tissue [4]. In particular, NIR light with a wavelength of 700-900 nm has optimal properties for measuring blood volume and oxygenation levels since most tissue – other than oxygenated and deoxygenated hemoglobin – absorb little light at these wavelengths. These qualities permit deep light penetration and backscattering from light sources such as white light, lasers, or LEDs, which can be measured by light detectors such as photomultiplier tubes or photodiodes.

In addition to its noninvasiveness, the low-cost and convenience of NIR imaging has been attracting much interest recently [5]. Some current NIR spectroscopy applications include brain functional imaging, breast cancer imaging, and muscle activity monitoring. For example, the finger pulse oximeter revolutionized hospital care with its ability to monitor arterial oxygen saturation and pulse rate while still being portable, noninvasive and capable of providing continuous real-time monitoring [6].

Given the success of contact model NIR devices, some current studies are moving toward remote sensing using NIR light. The goal of these studies is to combine the functionality of contact model NIR devices with the versatility of remote sensing, which might eliminate the subject's awareness of being tested altogether. One such contact model NIR idea of interest is brain oxygen monitoring in premature infants. The use of such a device could be extended to monitoring comatose or brain-injured patients and to measuring brain function.

Depending on the need, various NIR systems – including continuous-wave systems and time-resolution systems – can be used. A time-resolution system can measure the absorption coefficient and reduced scattering coefficient of tissue to find the absolute concentrations of deoxygenated hemoglobin (Hb) and oxygenated hemoglobin (HbO_2). From these two values, the absolute blood volume and oxygenation level can be determined [5]. However, this method relies on accurate measurement of photon arrival times that are backscattered from the tissue [7]. A much simpler method for NIR spectroscopy is a continuous-wave system, which merely emits a constant light at tissue and measures the backscattered light. An important limitation of a continuous-wave system is that it can only measure changes in Hb and HbO_2. Thus, only changes in blood volume and changes in oxygen levels can be measured [5].

All NIR methods use a NIR light source to illuminate tissue. The tissue surface reflects some of the incident light, while the rest enters the tissue. Light inside the tissue is either absorbed or scattered about until some of it reemerges. A light detector then measures this backscattered light. In the case of a continuous-wave system, the change in backscattered light is measured. Since Hb and HbO_2 are the greatest sources of absorption of the NIR light used, only these two are considered in the Beer-Lambert Law

$$I = GI_0 e^{-\left(\alpha_{Hb} C_{Hb} + \alpha_{HbO_2} C_{HbO_2}\right) L} \tag{1}$$

where I is the light intensity after absorption and backscattering; G is a constant attenuation; I_0 is the input light power; α_{Hb} and α_{HbO2} are the molar extinction coefficients of deoxygenated and oxygenated hemoglobin, respectively, and determine the amount of light absorbed; C_{Hb} and C_{HbO2} are the concentrations of deoxygenated and oxygenated hemoglobin, respectively; and L is the photon path length, which can be determined experimentally and varies based on the particular setup of the light source and detector [5]. Note that α_{Hb}, α_{HbO2}, and L are all functions of the light wavelength used. Water and other tissues have absorption coefficients that are orders of magnitude lower than Hb or HbO_2 and are not considered when dealing with NIR light [6].

A continuous-wave system needs at least two light sources of different wavelengths to operate. Let I'_{760} be the backscattered light intensity from a 760 nm light source at a predetermined "baseline state," and let C'_{Hb} and C'_{HbO2} be the unknown concentrations of deoxygenated and oxygenated hemoglobin concentrations at the baseline state. Then, for a light source of 760 nm, let the optical density (OD) be

$$
\begin{aligned}
OD_{760} &= \ln\left(\frac{I'_{760}}{I_{760}}\right) = \ln\left(\frac{GI_0 e^{-(\alpha_{Hb,760}\cdot C'_{Hb}+\alpha_{HbO_2,760}\cdot C'_{HbO_2})\cdot L_{760}}}{GI_0 e^{-(\alpha_{Hb,760}\cdot C_{Hb}+\alpha_{HbO_2,760}\cdot C_{HbO_2})\cdot L_{760}}}\right) \\
&= -\left(\alpha_{Hb,760}\cdot\left(C'_{Hb}-C_{Hb}\right)+\alpha_{HbO_2,760}\cdot\left(C'_{HbO_2}-C_{HbO_2}\right)\right)\cdot L_{760} \\
&= \left(\alpha_{Hb,760}\cdot\Delta C_{Hb}+\alpha_{HbO_2,760}\cdot\Delta C_{HbO_2}\right)\cdot L_{760}
\end{aligned}
\tag{2}
$$

where I_{760} is the backscattered light intensity from the 760 nm light source at some other state, and $\Delta C'_{Hb}$ and $\Delta C'_{HbO2}$ are the changes in concentrations of deoxygenated and oxygenated hemoglobin, respectively, from the baseline state. Similarly,

$$
OD_{830} = \ln\left(\frac{I'_{830}}{I_{830}}\right) = \left(\alpha_{Hb,830}\cdot\Delta C_{Hb}+\alpha_{HbO_2,830}\cdot\Delta C_{HbO_2}\right)\cdot L_{830} \tag{3}
$$

Thus, by using two wavelengths, the amount of change in Hb and HbO_2 from the baseline state can be determined:

$$
\Delta C_{Hb} = \frac{\alpha_{HbO_2,830}\cdot OD_{760}/L_{760}-\alpha_{HbO_2,760}\cdot OD_{830}/L_{830}}{\alpha_{Hb,760}\cdot\alpha_{HbO_2,830}-\alpha_{HbO_2,760}\cdot\alpha_{Hb,830}} \tag{4}
$$

$$
\Delta C_{HbO_2} = \frac{\alpha_{Hb,760}\cdot OD_{830}/L_{830}-\alpha_{Hb,830}\cdot OD_{760}/L_{760}}{\alpha_{Hb,760}\cdot\alpha_{HbO_2,830}-\alpha_{HbO_2,760}\cdot\alpha_{Hb,830}} \tag{5}
$$

Then, ΔBV, the change in blood volume, and ΔOXY, the change in oxygenated blood, can easily be calculated:

$$
\Delta BV = \Delta C_{Hb}+\Delta C_{HbO_2} \tag{6}
$$

$$
\Delta OXY = \Delta C_{HbO_2}-\Delta C_{Hb} \tag{7}
$$

If a third light source is used with a wavelength of 800 nm, at the isosbestic point [Fig. 1], then the change in blood volume, ΔBV, could be validated by OD_{800} alone since the molar extinction coefficients of Hb and HbO_2 are the same at this wavelength.

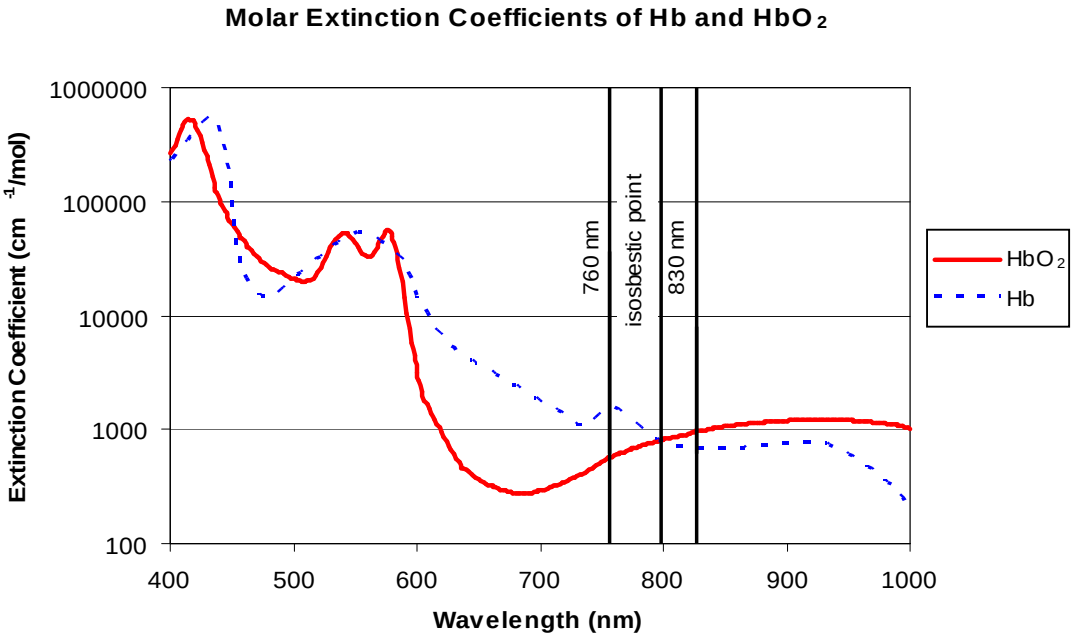

Figure 1: Molar extinction coefficients of Hb and HbO_2, as a function of wavelength, plotted on a logarithmic scale [8].

When monitoring brain health with a handheld remote sensing device, if a predetermined "healthy state" is used as the baseline state, then any changes or deviations in blood volume or blood oxygenation levels can be tracked to gauge brain health. Depending on the case, a tolerable range of fluctuation or deviation should be allowed, but once the change in blood volume or oxygenation levels from the predetermined healthy state exceeds these bounds, the device should alert the device operator. Similarly, when monitoring brain activity, a predetermined "resting state" could be used as the baseline state, and any changes in blood volume or blood oxygenation levels in a region of the brain could track brain function.

3. PROJECT GOALS

The goals of this project were to study the aspects of, analyze the feasibility of, and suggest a design for a handheld device capable of remotely sensing brain function by employing the basic building blocks of near-infrared technology. While it is currently the preferred procedure to attach a device to a patient's forehead for monitoring pulse rate and oxygen levels in the brain, the development of a remote handheld system would make it possible to spot check patients from a distance, without having to deal with the obtrusiveness of attaching probes. Current work suggests that affordable and safe handheld devices for contact systems can be built from inexpensive components such as LEDs and photodiodes. These components are small, safe, and operate at low voltage levels. A general system that can remotely measure brain function must include a light source and driver, a light detector, and a processing unit [Fig. 2].

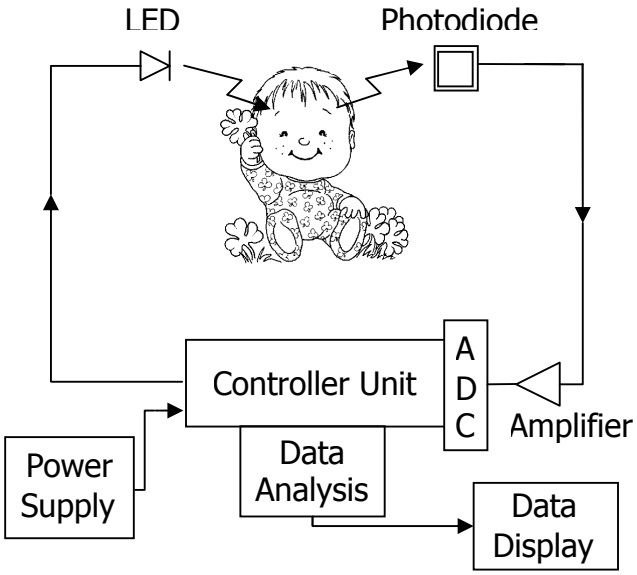

Figure 2: Block diagram of handheld device for remotely sensing brain function.

For this study, fundamental building blocks were selected to find the minimum requirements to build an effective, durable device at low cost. In addition to being small and cheap, LEDs generally have greater subject acceptance, and due to their more diffuse nature, the Food and Drug Administration does not limit their power when applied to humans. This provides LEDs with an advantage over lasers, which are subject to more federally determined limitations [5]. Due to their ease of use and versatility, photodiodes were selected as light detectors for this study. Unlike photomultiplier tubes, they do not require high voltages and can operate at light levels that would easily damage photomultiplier tubes due to the photomultiplier tubes' great sensitivity. Finally, an ordinary programmable microcontroller was selected to control the LED, sample the photodiode response with its built-in analog-to-digital converter (ADC), and process the sampled data. The object of much of this study was to determine the efficacy of these components when used together.

4. EXPERIMENTAL SETUP

Since the data measured with the device must be analyzed, there must be a way to sample, store, and interpret it. As with any digital device, an analog signal must be quantized in order to be sampled; the continuous range of the analog signal is broken up into a finite set of discrete values, to which the analog signal is mapped. In the case of the ADC selected for this study, a 0-5 V signal is uniformly quantized into a 10-bit value (1024 quantization levels), with 0 corresponding to 0 V and 1023 corresponding to 5 V. Thus, the precision of this particular ADC is $5/2^{10}$, or approximately 5 mV. The dominant problems with quantization are lack of precision and quantization error; this can be resolved by closely matching the quantized range to the range of the signal and by increasing the precision with more quantization levels. However, for a fixed range, fixed precision system, the only remedy is to use an analog system to fit the signal into the full dynamic range of the ADC.

For example, if the component of the desired signal is too small, the discrete nature of the quantization levels may not be able to detect the differences of the actual continuous, infinite precision values. In this situation, all that can be done is to extract the desired signal and amplify it to better fit the dynamic range of the ADC so that more accurate sampling can be done [Fig. 3].

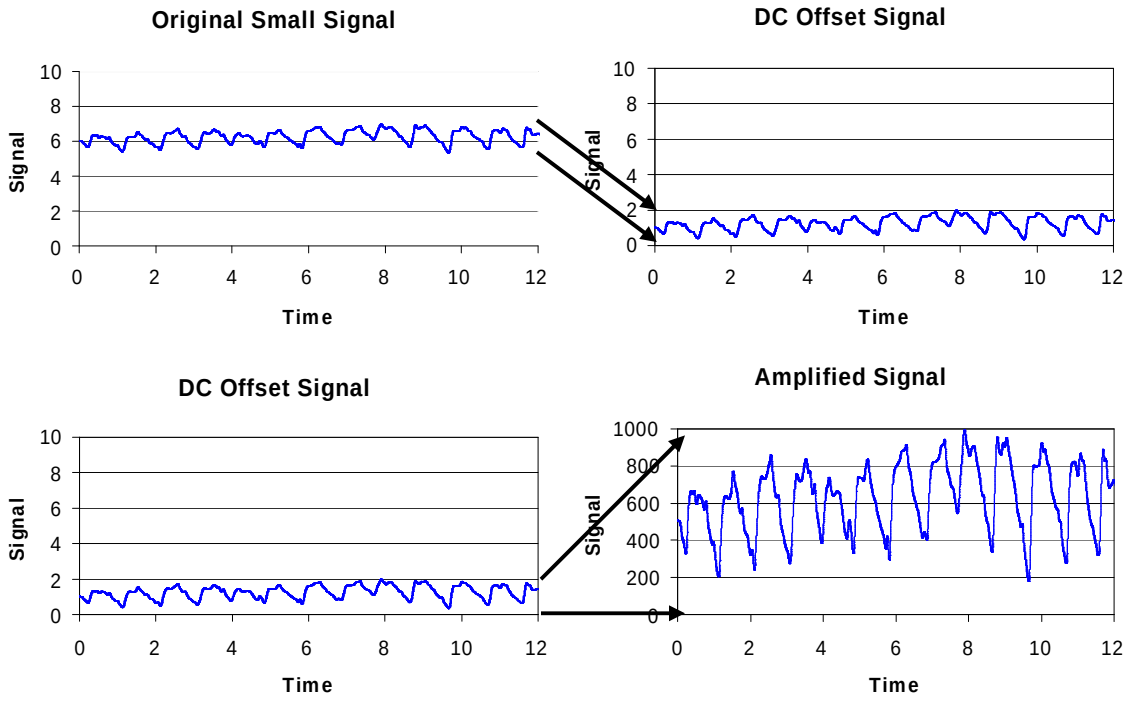

Figure 3: The original small signal's DC component is first reduced, and then the signal is amplified to better match the ADC's full dynamic range.

This must be done with analog components, which are continuous by nature. The circuit built for this project performs this exact function by first introducing a DC offset to crop off the DC component. Then the signal is amplified to better fit the dynamic range of the ADC before it is sampled. Nonetheless, due to the discrete nature of quantization, noise is introduced. Even for an ideal *Q*-bit analog to digital converter, the theoretical Signal to Noise Ratio (SNR) is [9]:

$$SNR_{ADC} = (1.763 + 6.02 \cdot Q) \text{ dB}. \tag{8}$$

Thus, for the 10-bit ADC used, the theoretical maximum SNR is 62.0 dB, which despite being an upper bound, is a very impractical bound; noise introduced from numerous other sources greatly degrades the signal, and the SNR is not limited by the ADC quantization.

Much of the circuit used was built from op-amps, a fundamental element of circuits. These conceptually simple devices have a wide variety of uses ranging from logic comparators to amplifiers to integrators. The first step necessary to take advantage of the ADC's full dynamic range is to remove the DC component so that only the

component of the signal of interest is amplified. This was accomplished by subtracting a DC offset ranging from 0 V to +5 V that could be adjusted by a potentiometer. Next, another op-amp configuration amplified the signal with a gain up to 100, adjusted by a 100 kΩ potentiometer [Fig. 5].

Depending on the need, the Motorola 9S12C32 microcontroller was programmed to sample at certain frequencies or to drive two LEDs (760 and 830 nm). For simplicity, the sampled data was stored and analyzed offline. As the project progressed, the necessity of using higher quality components became apparent. A Wratten gelatin light filter #89B placed over the photodiode filtered out undesired wavelengths [Fig. 4], and greatly aided in reducing noise since photodiodes respond to a wide range of wavelengths.

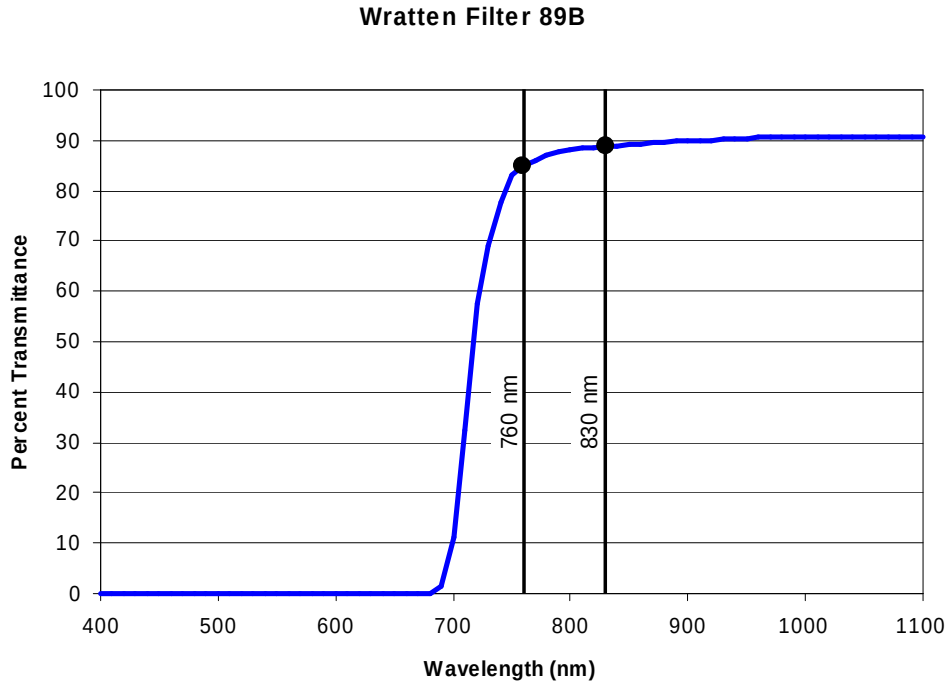

Figure 4: Wratten filter #89B's percent transmission curve as a function of wavelength [10]. The filter minimally affects the two LEDs used while lower frequencies (visible light) are blocked.

Higher precision, lower noise OP27 op-amps replaced the more standard op-amps originally used (80 nV compared to 1000 nV peak-to-peak noise in the 0.1 to 10 Hz range), and the OPT101 photodiode was replaced by the FDS1010 Si photodiode (9.7 × 9.7 mm), which has 18 times the surface area for collecting light. When the passive FDS1010 Si photodiode was implemented, the leads were fed through a difference amplifier with a pre-amp gain of 10 [Fig. 5]. Furthermore, a first order RC low-pass filter was added to reduce high frequency noise in the signal, such as the 60 Hz AC component in power from wall outlets and room light.

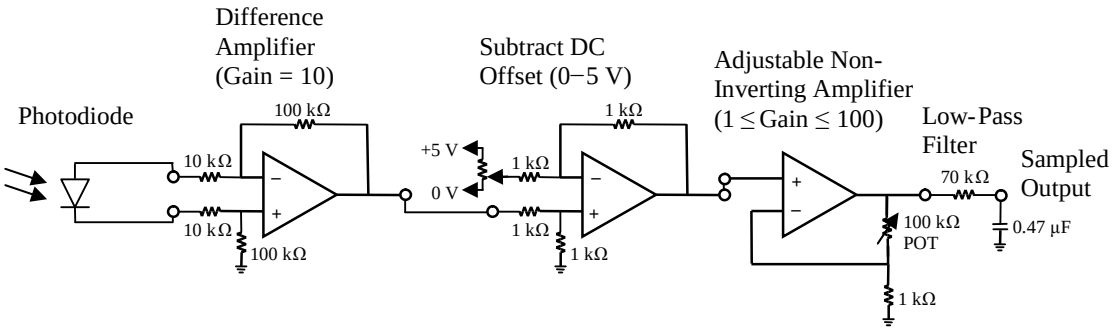

Figure 5: Circuit diagram showing op-amps (±12 V power supply) used to extract signal throughout this project.

5. EXPERIMENTAL RESULTS

Two aspects of remote sensing were explored: 1) an attempt to find the arterial pulse and 2) an attempt to track changes in light absorption of an ink test via remote sensing. The former illustrates the need for finding the pulse rate of the subject, and the latter simulates changes in concentrations of Hb or HbO_2.

5.1 Arterial Pulse

Starting with the contact model, where both the LED and photodiode are in contact with the subject, the arterial pulse is easily extracted. An LED shines light through the subject's thumbnail, and a photodiode measures the transmitted light through the thumb. The amount of blood at the tip of the thumb fluctuates in relation to the arterial pulse. This rhythmic fluctuation is the basis for determining the pulse rate. As the blood volume increases, the amount of light transmitted decreases due to the greater absorption of NIR light by Hb and HbO_2. The raw data was run through a digital 4th order Butterworth low-pass filter to clean up the signal. Using a Discrete Fourier Transform, the pulse rate of the subject (in this case, the author) was determined over a sampling duration of 30 seconds to within 2 beats per minute to be 74 beats per minute [Fig. 6]. This was validated by counting the pulse rate on the wrist. In this setup, the presence of room light was not a problem since it merely contributed to the transmitted light and was still modulated by the arterial pulse.

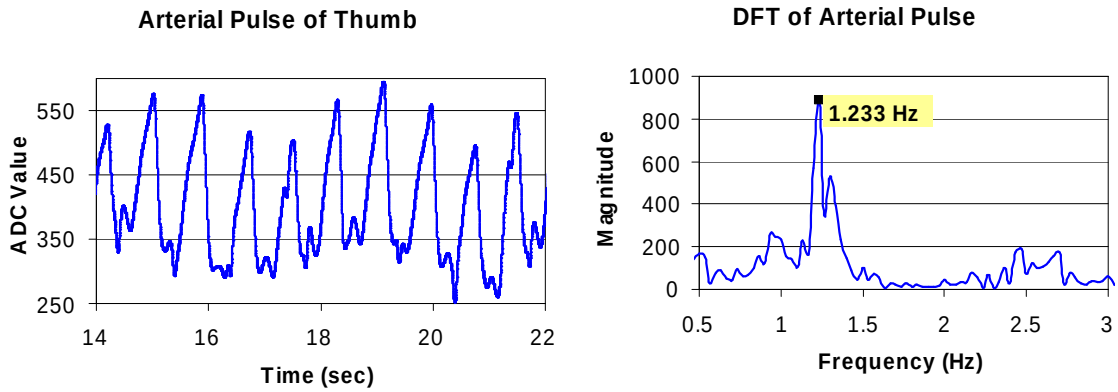

Figure 6: Measured transmitted light (left) through thumb with a contact setup, and its frequency spectrum (right), showing a pulse rate of approximately 74 beats per minute.

After showing that the contact model could be reproduced, a small-scale setup for remote sensing [Fig. 7] was built to test the feasibility of remote sensing. Neither the LEDs nor the photodiode were in direct contact with the tissue; instead, they were fixed a small distance from the tissue, as a step toward achieving distances more likely for remote sensing. A barrier prevents light from the LED from directly illuminating the photodiode and reduces reflected light from the surface of the tissue. Over the course of hundreds of tests, it was determined that extracting the arterial pulse from the measured backscattered light required: 1) higher quality op-amps, 2) low-noise OP27 chips, 3) the addition of a light filter, 4) the change to a greater surface area FDS1010 Si photodiode, and 5) the reduction of incident room light. Even with these enhancements, extracting the arterial pulse from this scaled down setup of remote sensing proved difficult because of the small signal and noise.

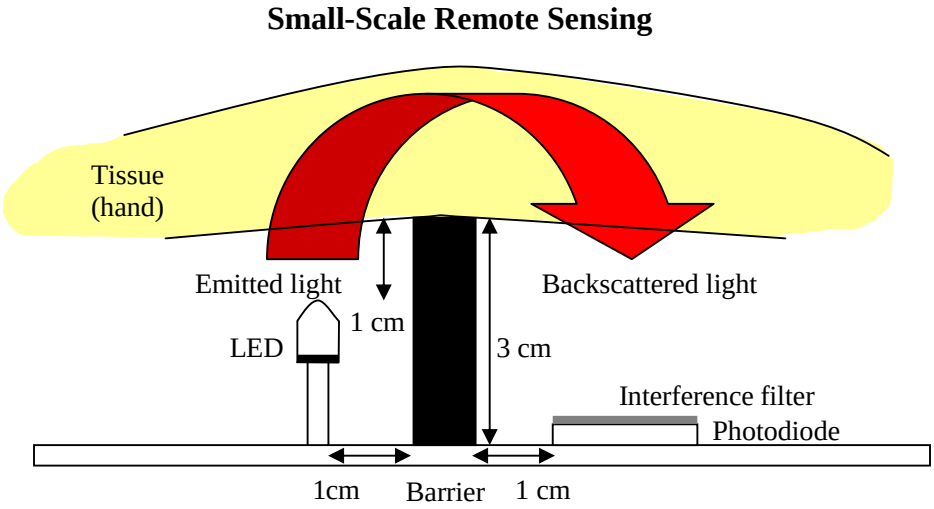

Figure 7: Small-scale remote sensing setup with neither the LED nor the photodiode in contact with the tissue.

In order to utilize Equations 2-7, which can determine the change in blood volume or oxygen levels in tissue, two LEDs are necessary. To monitor the change in the intensity of backscattered light from each wavelength, the two LEDs cannot emit light simultaneously; instead, they are time-shared and alternately emit light at a rate much greater than the rate of change of the parameter of interest. For finding the arterial pulse, the microcontroller controls a 760 nm and an 830 nm LED by flashing them alternately for a 100 ms duration each in the small-scale remote sensing setup. The average backscattered light intensity is different for each LED, but the arterial pulse modulates the backscattered light for both wavelengths [Fig. 8].

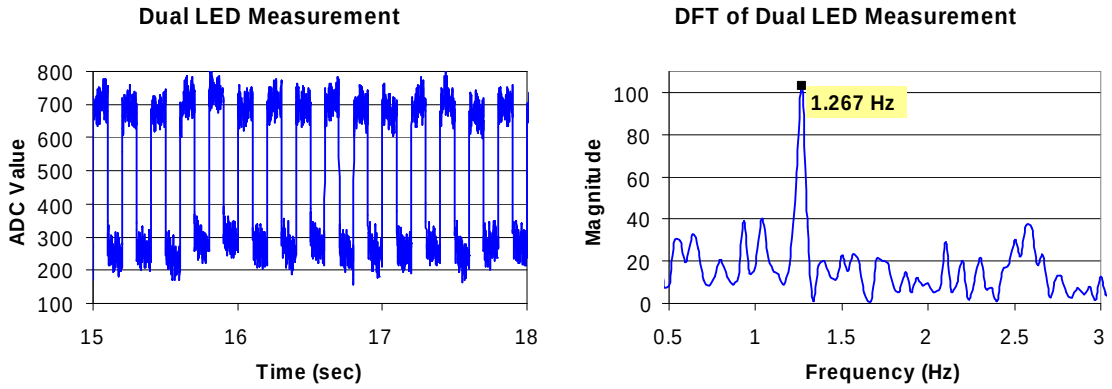

Figure 8: Measured backscattered light (left) from the author's hand with the small-scale remote sensing setup and two time-shared LEDs; the frequency spectrum of the backscattered light (right), showing a pulse rate of approximately 76 beats per minute modulating the backscattered light.

5.2 Ink Tests

In addition to remotely sensing a subject's pulse rate, another goal for the handheld device is to monitor changes in blood volume and oxygen levels in the brain. To achieve this, the handheld device must be sensitive to changes in concentrations of Hb and HbO_2, which can be simulated by an ink test. An intravenous fat emulsion solution in a large plastic beaker provided the basis for a scattering medium that allows incident light to be backscattered [Fig. 9]; a photodiode collects the backscattered light. Ink is slowly added to change the amount of light absorption, just as a change in the concentration of Hb or HbO_2 would change the amount of light absorption. Various setups were tested, with varying distances between the beaker and the light barrier, with both the 760 and 830 nm LEDs, and with and without the presence of room light.

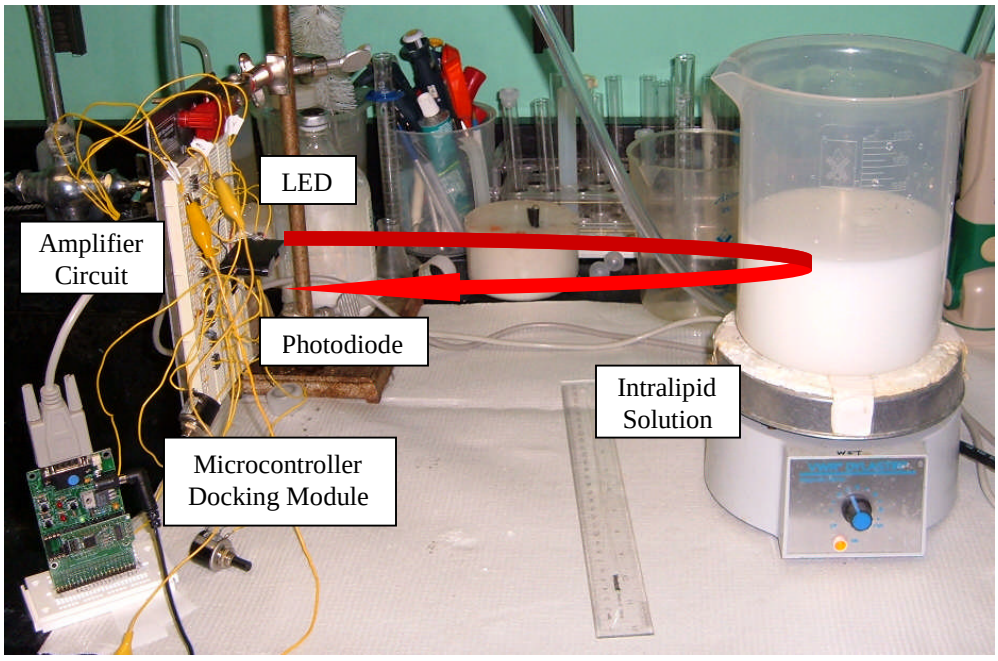

Figure 9: Remote sensing ink test setup with handheld device components.

The photodiode sampling for the ink tests was different from the arterial pulse tests since no specific frequencies were being tested for; only the general trend over a long period of time was being tested for. For example, with the beaker 10 cm from the LED, no room light, and a 760 nm LED, the measured backscattered light shows clear decreases as 1 mL of ink is added approximately every minute [Fig. 10] to the 1 L of intralipid solution. Each vertical division marks the addition of ink to the intralipid solution, which increases the absorption of light and reduces the amount of backscattered light. Before each addition of ink, the signal is stabilized and shows little change; as ink is added, a press of a button on the microcontroller's docking module records the action of adding ink.

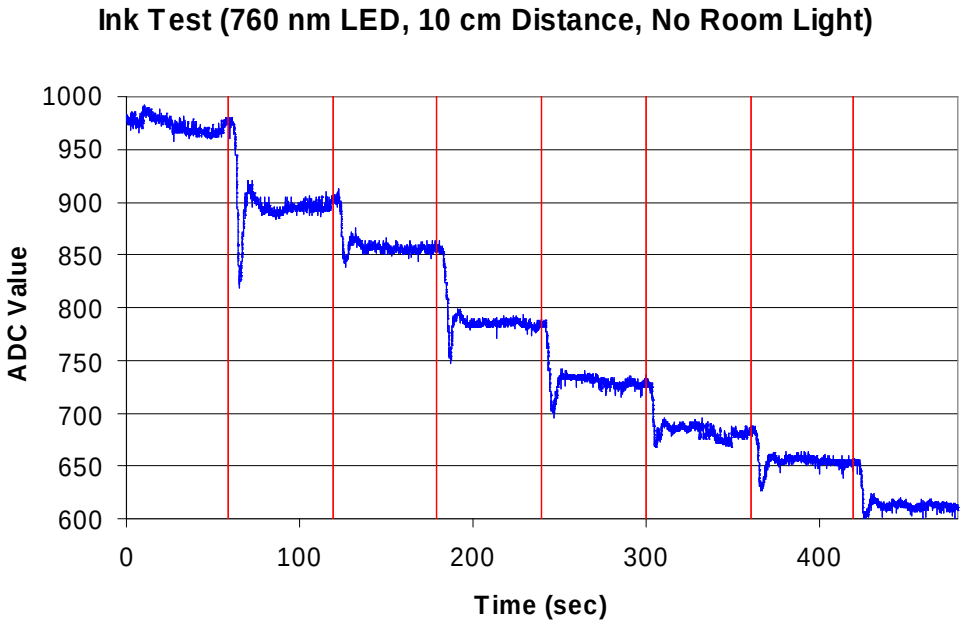

Figure 10: Ink test with a 760 nm LED at a distance of 10 cm from the beaker with no room light. The backscattered light intensity drops with each addition of ink, marked by the red vertical lines.

While the 760 nm and 830 nm LEDs showed similar effectiveness in the different ink test configurations, the distance between the beaker and LED and whether room light was present greatly affected the ability of the system built to determine the decrease in backscattered light when ink was added. Adding room light to the configuration used to obtain Figure 10 injects significantly more noise to the raw data and requires twice the gain to extract a similar signal even with the Wratten light filter; changing the distance from 10 cm to 30 cm without room light also significantly weakens the signal and requires five times the gain to extract the signal [Fig. 11]. At a distance of 30 cm between the beaker and the LED, with the presence of room light, the downward trend in backscattered light due to the addition of ink could no longer be measured; at a distance of 60 cm without the presence of room light, this could no longer be accomplished.

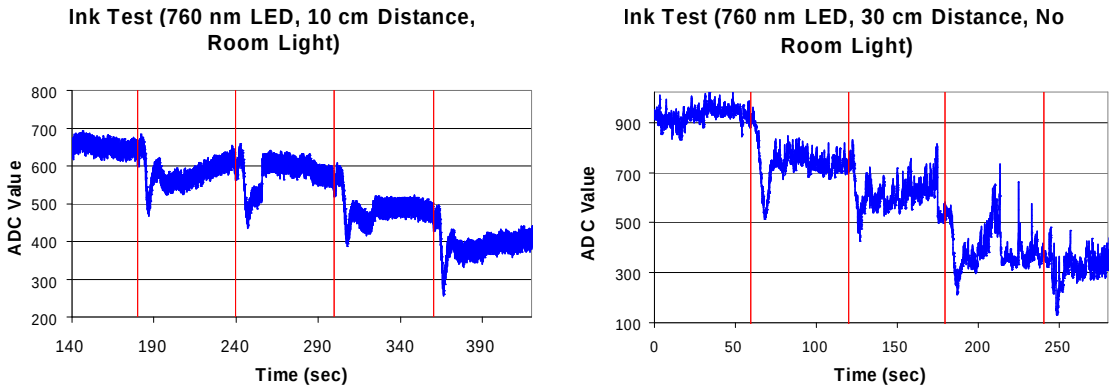

Figure 11: Ink tests with room light present (left) and at a greater distance (right), show a general degradation of the signal but still demonstrate a response to ink being added.

6. DISCUSSION

While the basic components used in this study have shown some success in finding the pulse rate and tracking changes in absorption values with remote sensing, there is much left to be desired. Even small-scale remote sensing detection of the arterial pulse required careful shielding from room light. The ink tests also showed that distance between the subject and device and the presence of room light greatly affected the ability to accurately track changes in light absorption. More realistic remote sensing scenarios for such a device might require distances of up to several feet between the subject and the device in the presence of room light. High gain circuits further exacerbate the problem of high sensitivity, requiring an extraordinarily stationary subject and operator. Furthermore, room light injects significantly greater noise, especially as the signal of interest weakens with distance, with a large 60 Hz component that must be diminished by a low-pass filter.

When two LEDs of different wavelengths are time-shared – as they must be to monitor changes in blood volume or oxygen levels – the backscattered light intensity from the two different wavelengths will differ. Therefore, if the same gain circuit amplifies the signal from both LEDs, the difference in backscattered light intensity between the two LEDs will be amplified, and both signals may not fit in the ADC's range. To solve this problem, the gain could be lowered to accommodate for the different backscattered light intensities. However, this change causes the signal from each LED to be smaller too. Either the LED power should be adjusted so that the backscattered light intensities from the two LEDs are very close in value, or, better yet, each wavelength should have its own DC offset and gain circuit that the microcontroller could sample at the appropriate time.

Since the variable volume of blood due to the arterial pulse only accounts for 5% of the total blood volume in tissue [6], finding the arterial pulse from backscattered light would be much more challenging than tracking general changes in the overall trend of blood volume and oxygen levels in the brain. The ink tests demonstrated the *qualitative* ability of the handheld device components to track changes in absorption values; however, a *quantitative* indicator is needed. To accomplish this, effective blood tests must closely

simulate the scattering and absorption coefficients of human tissue and quantitatively calibrate a system with known changes in the absorption coefficients at the wavelengths used.

At the end of the project, the issue of bandwidth and shot noise arose. Shot noise, which is inherent to photodiodes, is proportionally related to the square root of the bandwidth of the photodiode [5]. Therefore, in order to limit shot noise, the bandwidth of the entire system should be limited to the signal of interest, which, at its greatest frequency, would probably not exceed a few hertz. Some tests showed that simply adding a capacitor in parallel to the negative feedback resistor of the difference amplifier could significantly reduce shot noise. The system originally built did not take shot noise into consideration and had a bandwidth of approximately 600 Hz. Adding a 47 µF capacitor that would reduce the bandwidth to approximately 1 Hz still allows for the detection of the arterial pulse but reduces shot noise by almost a factor of 25 [Fig. 12]. Further tests would be required to determine the effect of shot noise.

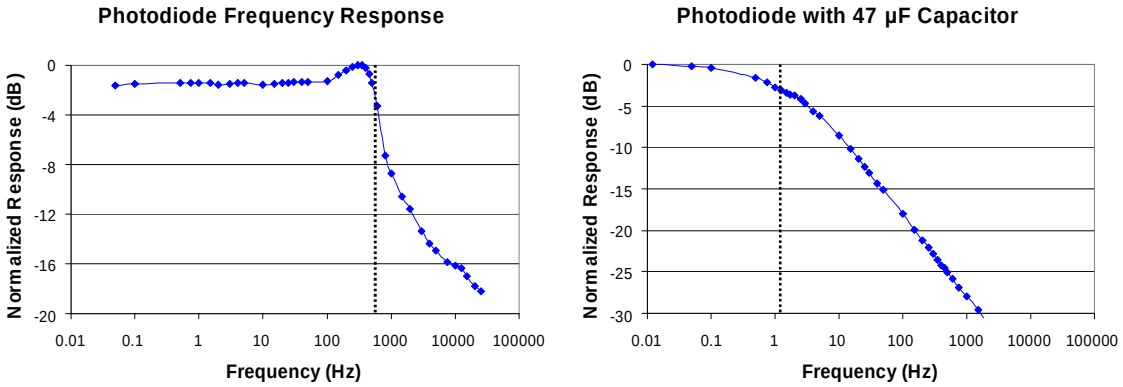

Figure 12: System built for this study (left) with a bandwidth of about 600 Hz that can be improved by filtering noise early on by limiting the bandwidth.

7. RECOMMENDATIONS

Due to the weak signal and the high level of noise inherently present in remote sensing, sources of noise should be targeted aggressively – clean power sources and low-noise components are essential. The Wratten gelatin filter used was a good start for reducing unwanted room light, but a better, more expensive solution would be a narrow bandpass lens coating that would exclusively pass the wavelengths of the LEDs. Furthermore, the gelatin filter is sensitive to cleaning, and a handheld remote sensing device should be durable and hardy, considering the more rigorous uses intended outside of the lab.

A weak signal entails the need for a high gain circuit, which also unfortunately amplifies sampled noise. Thus, various measures could be enacted to obtain a stronger signal, such as reducing ambient lighting, increasing the LEDs' power, minimizing the distance between the device and the subject, and maximizing the collecting area of the backscattered light. An array or multiple LEDs or photodiodes could be explored as a

possibility to increase the effective illumination intensity or effective collecting area of the handheld device. A simple way to increase the effective collecting area of the photodiode would be to use a large lens to collect light and focus it onto the photodiode.

Since the change in blood volume or oxygen level in the brain is determined by the change of measured backscattered light intensity, care must be taken to limit any movement of the subject or the operator of the handheld remote sensing device. Slight angular movements of either the device or the subject could change the amount of incident light intensity from the LED onto the subject or backscattered light intensity from the subject onto the photodiode [Fig. 13]. Additionally, if the LED is considered as a point source, the light intensity varies inversely with the square of the distance from the source to the subject. In the presence of room light, even a passerby's shadow would dramatically change the readings due to the high gain necessary to accurately sample the photodiode.

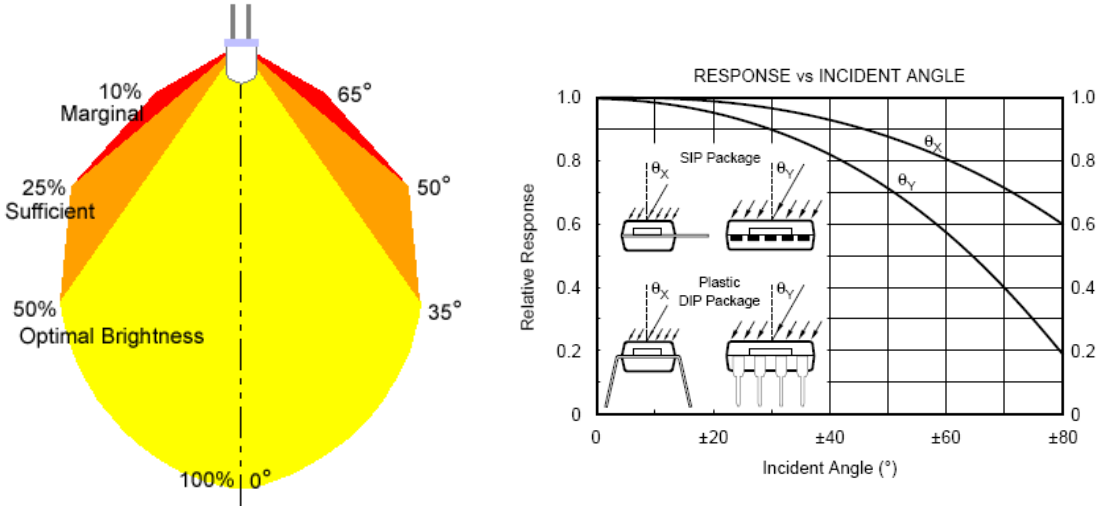

Figure 13: Relative brightness of an LED (left) [11] and relative response of a photodiode (right) [12], based on the incident angle.

The microcontroller selected for this study would not have any trouble performing the calculations demanded by Equations 2-7 to monitor changes in blood volume or oxygen levels in the brain. Its I/O ports could digitally display real-time data on an LCD, light up various indicator LEDs as to the subject's health, read user inputs, and sound audible warning alarms. Moreover, instead of the analog potentiometer knobs used for this study, the microcontroller could control more precise digital potentiometers; the gain and DC offset could be adjusted automatically based on the need to keep the signal in the ADC's range and to take advantage of the ADC's full dynamic range. All changes to the potentiometers would be recorded and taken into consideration in the microcontroller's calculations. Nonetheless, a Digital Signal Processor may be desired for its generally greater computational power, ADC range, and quantization precision.

8. CONCLUSION

This study has shown the feasibility, challenges, and limitations of remote sensing with an LED, photodiode, and microcontroller. Together, these components could provide a compact, durable, and effective handheld device if a consistent and stable signal can be extracted. Small-scale remote sensing tests have demonstrated the ability to determine the pulse rate of a subject and changes in blood volume or oxygen levels in experiments designed to simulate measuring brain health or brain function. Further study that implements some of the recommendations of this paper could very possibly bring remote sensing in the real world to reality. Even so, the exciting possibility of remote sensing with a handheld device requires much additional work before it can be an effective technology.

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