At-home Sleep Apnea Test Using Pulse Oximetry

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Abstract

Many patients who suffer from sleep apnea do not get proper diagnosis due to the discomfort and complicacy of the current diagnosis systems. The main focus of the project is to provide a solution for this problem by designing a home sleep apnea test. A patient's blood oxygen saturation level goes down during the apnea attack and this phenomenon can be detected using a pulse oximeter. Therefore a simple pulse oximeter is designed to provide data for the software that detects the risk of OSA. The software designed cannot detect sleep apnea with absolute certainty, but can provide with accurate risk warning for a patient who might be suffering. The output of the designed system is given by a software designed using MATLAB. Certain limitations exist in the development process and those are considered negligible for the scope of the project. However the solution of the limitations and alternate approach of the assumptions are discussed in details.

Key Words: Sleep Apnea, OSA, Approximate Entropy, Pulse Oximeter, Sleep Study, Oxygen Saturation

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Nomenclature

OSA – Obstructive Sleep Apnea

Apnea Episode – pauses in breathing for a OSA positive patient

Apnea Attack – see Apnea Episode

AHI – apnea-hypopnea index

SaO_2 – Arterial Oxygen Saturation

Hb – deoxygenated hemoglobin

HbO_2 – oxygenated hemoglobin

ApEn – Approximate Entropy

IR – Infrared

LED – Light Emitting Diode

1. Introduction

1.1 Background

Obstructive sleep apnea is a breathing disorder that is caused when the tissues or muscles of the upper airway collapse blocking the inflow of air temporarily during sleep. It can be caused by various reasons such as obesity, tonsil and adenoid enlargement, natural narrowing of airway, nasal obstruction, alcohol or sedative use etc. Sometimes the condition is hereditary. Sleep apnea can be dangerous because a person suffering from OSA often does not realize the condition because the episodes occur during sleep. The most common sign of sleep apnea is snoring, which can also be caused by various reasons. However sleep apnea snoring is different from other forms, as it is often interrupted by long silent period when the patient does not breath followed by gasping to restart breathing. Other OSA symptoms include daytime fatigue, inappropriate falling asleep, memory difficulties, headaches, personality changes, poor concentration, possible depression, impaired emotional functioning etc. [1]

Besides the aforementioned discomforts sleep apnea can also cause other complications. Stroke, high blood pressure, heart failure, hallucinations etc are common for an undiagnosed OSA patient. Moreover, many cases exist where a person with sleep apnea fell asleep during job or driving causing risky accidents [1]. It is estimated that one in four middle aged men and one in ten middle aged women suffer from sleep apnea, where only 10-20% is diagnosed. [4]

AHI is the tool that is currently used to detect sleep apnea. It measures the number of times of apnea attack occurring per hour during sleep. Two major testing systems exist to diagnose sleep apnea. One is a nocturnal polysomnography and the other one is an at home sleep apnea test. The overnight polysomnography monitors ECG, EOG, EMG, thoracic airflow, pulse oximetry, continuous video monitoring etc. [2]. The at home test is also similarly extensive and a medical professional is needed to set up the system.

As sleep apnea restricts the breathing rate, it causes drop in blood oxygen saturation level. To understand the theories of blood oxygen saturation and pulse oximetry, it is crucial to understand how Hemoglobin plays the role of oxygen transporter in the body. One red blood cell contains approximately 265 molecules of hemoglobin. One hemoglobin molecule contains four heme and four globin units and each of a heme-globin unit can carry one oxygen molecule. Therefore one hemoglobin molecule can carry four molecules of oxygen. [3]

A property of hemoglobin is that it changes color when oxygenated. Oxygenated hemoglobin is bright red and deoxygenated hemoglobin is dark red, therefore the light absorbance of oxygenated and deoxygenated hemoglobin is different. Also, the absorbance is different for light of two different wavelengths (figure 4). These principles are used to calculate blood oxygen saturation or SaO_2 and will be discussed further in section 3.1.

For a healthy human being, blood oxygen saturation or SaO_2 should be 90% - 100% [6]. A patient suffering from sleep apnea has overall low SaO_2 level. Moreover, studies show that SaO_2 level can drop significantly during an apnea episode. Figure 1 shows the pattern of the overnight blood oxygen saturation for a healthy case (a) and a case of sleep apnea (b). Case b exhibits the dips in the time series showing the apnea attacks during the study period. [5]
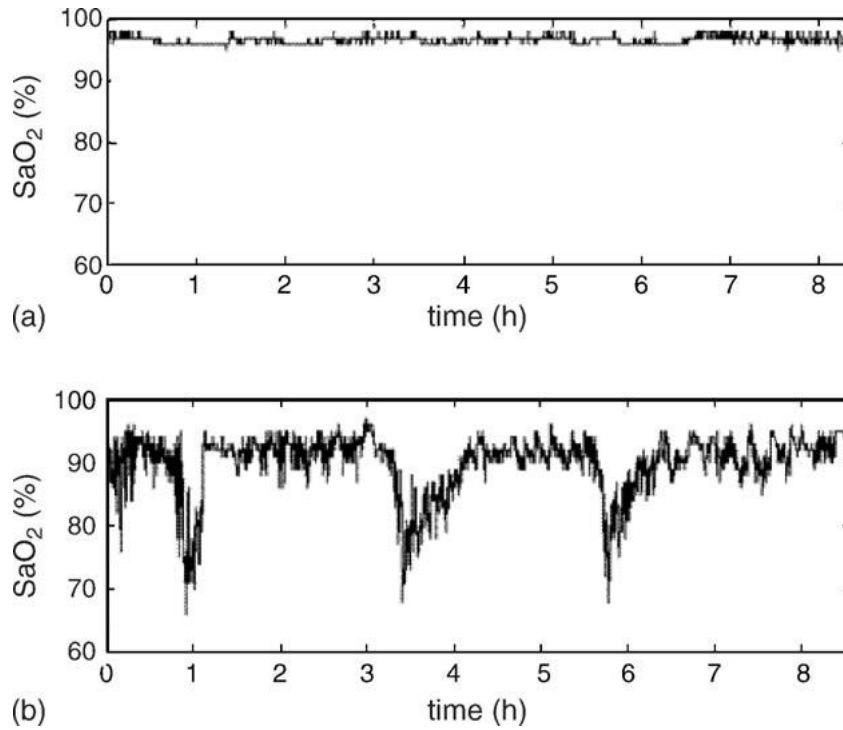
Figure 1: SaO$_2$ pattern for overnight study for a (a) normal case, (b) case of sleep apnea [5]

1.2 Objective

As mentioned in the previous section, tests are available to diagnose sleep apnea. However, these tests are very extensive and uncomfortable for the patient. Also, one or a few nights of study might not reflect what happens during sleep for a patient in the regular basis. Therefore, a necessity appears for a simpler test that can efficiently predict a patient's risk level for sleep apnea.

The objective of this project is to create a pulse oximeter which can perform overnight monitoring of oxygen level in blood and use that result to asses a patient's risk of having OSA.

1.3 Scope of the project

The device designed for this project has the ability to calculate blood oxygen saturation and heart rate of the patient. The heart rate is vital information for someone who is prone to sleep apnea.

The blood oxygen saturation is an indicator of sleep apnea. The software has the capability of calculating the entropy of the SaO2 which then indicates if the patient is high risk for sleep apnea.

As this device is only designed as a preliminary prototype it includes a few assumptions and limitations in both the hardware and software design. These assumptions will be described in the course of explanation for the design procedure.

The device designed in this project is not a replacement of the existing methods of detecting sleep apnea, but is designed for individuals who have the symptoms of sleep apnea or other reasons to be concerned about their blood oxygen saturation. But further improvement in the accuracy of the designed system can show promising result in sleep apnea detection.

2. Literature Review

The main focus of the literature search of this project was based on the following areas

1. LED selection for finger clip
2. Circuit design for signal manipulation
3. Calculation of SaO_2
4. Find connection between SaO_2 and sleep apnea

The description for the logic behind the LED selection, the circuit design and the SaO_2 calculation procedure is described in section 3, 'problem statement and solution methodology'.

Research is being done that shows promising results that relates pulse oximetry and sleep apnea. Poincare analysis [19], standard deviation calculation and entropy analysis [5] are the most common in the analysis procedure. However, the most promising and clear indication of sleep apnea can be noticed in the approximate entropy analysis. One of the studies that had been conducted in this area includes a group of 187 subjects of whom, 111 are diagnosed as sleep apnea positive and 76 diagnosed as sleep apnea negative. ApEn calculation (appendix A) was then performed for each subject on an overnight sleep study data of oxygen saturation. Figure 2 shows the result obtained in this study, and the optimum threshold for OSA positive group was found to be 0.77 [5]. Therefore, from this study, it can be safely concluded that result above 0.77 shows high risk of sleep apnea.
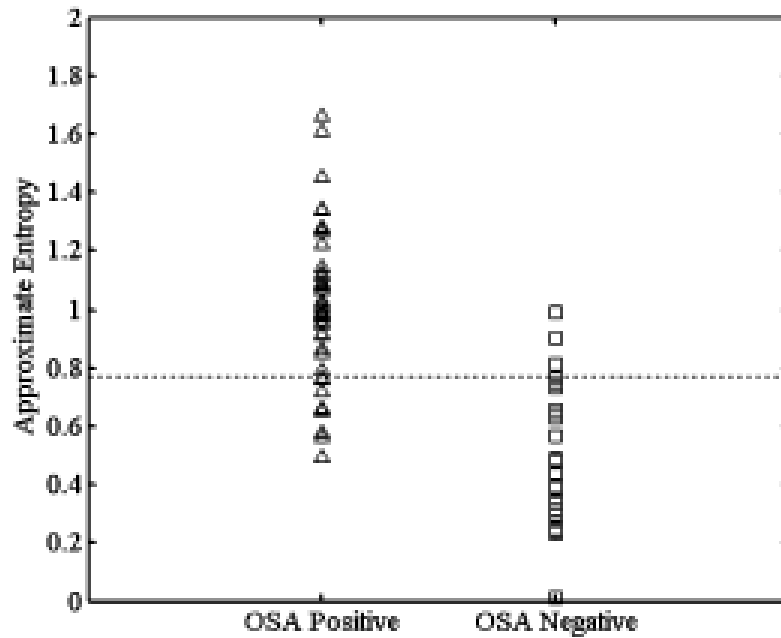
Figure 2: Approximate entropy result pattern showing that people with higher that 0.8 ApEn is more likely to have sleep apnea [5]

This result can be used to predict the risk of OSA but cannot detect positive or negative sleep apnea as a significant number of cases for sleep apnea positive lie beneath the threshold value and a few OSA negative cases exist above the threshold value.

3. Problem Statement and Solution Methodology

According to the project proposal, the final device should be able to perform an overnight study that measures the patient's SaO_2 level for certain time intervals and then provide risk assessment for the possibility of OSA. The primary element of this project is to understand how SaO_2 is calculated and develop a calculation algorithm. The next aspect is to design the sensor to acquire the necessary signal and manipulate the signal to a usable format on which certain calculation should be applied to achieve the final result. The main steps of the solution method can be represented graphically with the following diagram.
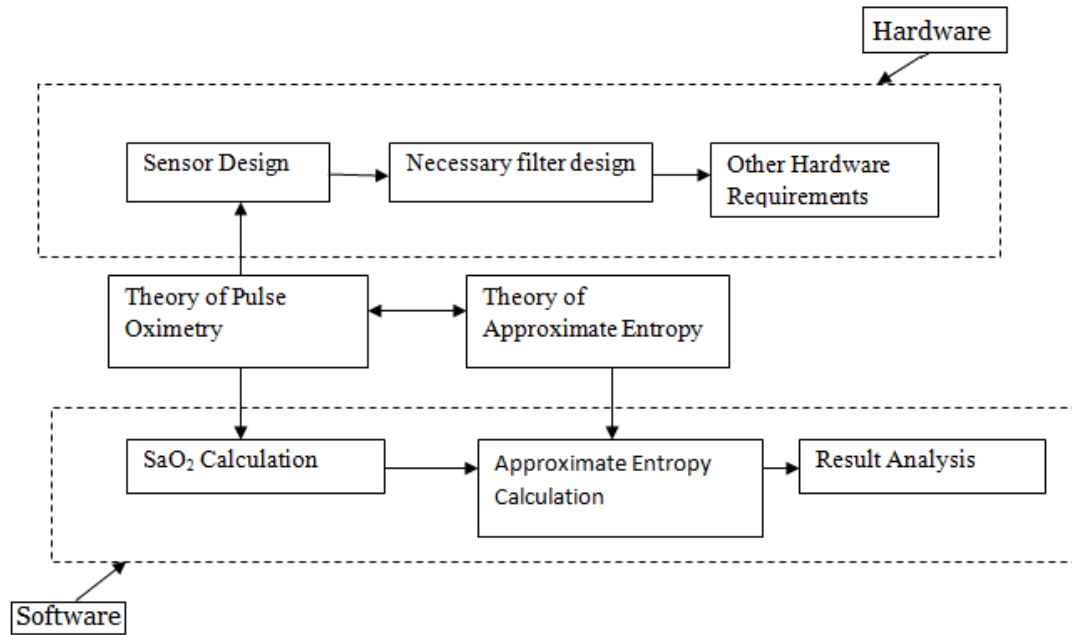
Figure 3: Project Solution Methodology

3.1 Theory of Pulse Oximetry

The theory of pulse oximetry is based on the light absorbance properties of oxygenated and deoxygenated hemoglobin. Figure 4 shows that oxygenated hemoglobin (HbO_2) absorbs more infrared light and deoxygenated hemoglobin (Hb) absorbs more red light.
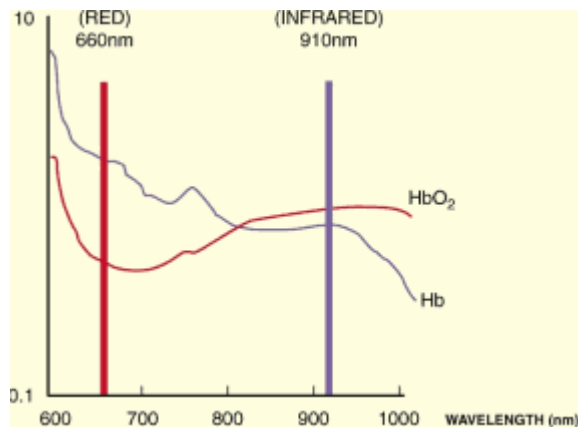
Figure 4: absorbance pattern of oxygenated and deoxygenated hemoglobin for different wavelength (Absorbance vs. Wavelength) [7]

In pulse oximetry, two LEDs are used which are usually a pair of red and infra red lights. The light from these LEDs are passed through a comparative translucent part of the body usually earlobe, finger or toe. [7]

Two methods exist to design the sensor that can detect the blood absorption: transmission/forward scattered model and reflection/ backscattered model. In the forward scattered model, the LEDs and the photodiode are situated in the different side of the body and in the backscattered model, the LEDs and the photodiode exist on the same side of the body. [8]

During systole, the heart pumps out the blood therefore, the arteries contain more blood than during diastole. This causes existence of more light absorbing components (Hb and HbO_2) to exist in blood during systole. Due to the pulsatile nature of blood, the output of the photodetector contains an AC component. It also contains a large DC component caused by constant light absorption of venous blood, tissue, bone etc. Figure 5 shows the shape of the function expected at the output of the photodetector.
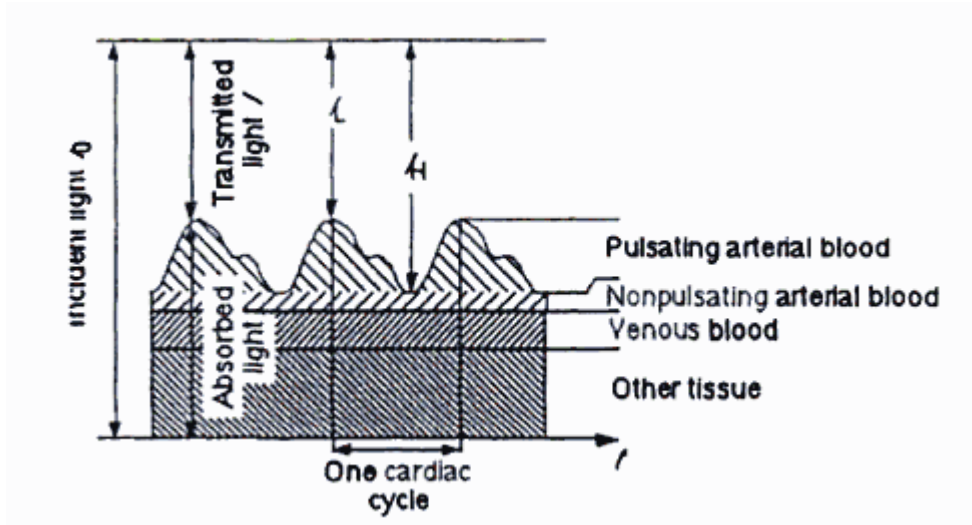
Figure 5: Expected output of the pulse oximeter circuit [3]

Using Beer-Lambert`s law and this absorbance properties of blood, the following equation has been developed to calculate SaO₂ [3].

$$SaO2 = \frac{a_{Hbr} - a_{Hbir}\,R}{a_{Hbr} - a_{Hbo\,2r} + (a_{Hbo\,2ir} - a_{Hbir})R} \; ----(1)$$

Where,

a_{Hbr} = Extinction coefficient of Red light in de-oxygenated hemoglobin

a_{Hbir} = Extinction coefficient of Infra Red light in de-oxygenated hemoglobin

a_{Hbo2r} = Extinction coefficient of Red light in oxygenated hemoglobin

a_{Hbo2ir} = Extinction coefficient of Infra Red light in oxygenated hemoglobin

R = ratio of the output intensity (logarithmic)

However, signal manipulation is easier when the output of the photodetector is transformed to voltage. Therefore the ratio R is calculated in terms of voltage, not intensity.

$$R = \frac{\ln\left(\frac{VmaxR}{VminR}\right)}{\ln\left(\frac{VmaxIR}{VminIR}\right)} ----(2)$$

VmaxR = maximum peak due to red light during voltage as output

VmaxIR = maximum peak due to infra red light during voltage as output

VminR = minimum peak due to red light during voltage as output

VminIR = minimum peak due to infra red light during voltage as output

Table 1: Extinction coefficient table [3]

Light(wavelength)	Extinction coefficient of Hb Lmmol^{-1}cm^{-1}	Extinction coefficient of HbO2 Lmmol^{-1}cm^{-1}
Red - 640nm	0.81	0.08
Infra Red – 960nm	0.18	0.29

Using the values from table 1 and equation 1, we can find the following equation to calculate SaO_2.

$$SaO2 = \frac{0.81 - 0.18R}{0.73 + 0.11R} --- (3)$$

Note that the output signal for both red and infrared LED must be normalized before the peak detection. [3]

3.2 Heart Rate Calculation

Heart rate is a by product of the signal acquired from the output of the photodetector. Figure 5 shows that the distance between two peaks of the output represents one cardiac cycle. If the time required in one cardiac cycle is 'T' seconds (distance between the two peaks),

Heart Rate = 60/T beats per minute. -----(4)

3.3 Sensor Design

A forward scattered pulse oximeter sensor specially made to use on the finger is designed for the purpose of this project. In this type of design, the infrared and the red led located on the same side of the finger.

Infrared and Red LEDs

Finger
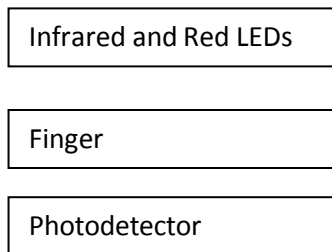
Photodetector

Figure 6: Location of the LEDs with respect to the finger

Photodetector provides output as current proportional to the emitted intensity. A simple current to voltage converted was designed to provide a voltage input for the filtering circuitry.
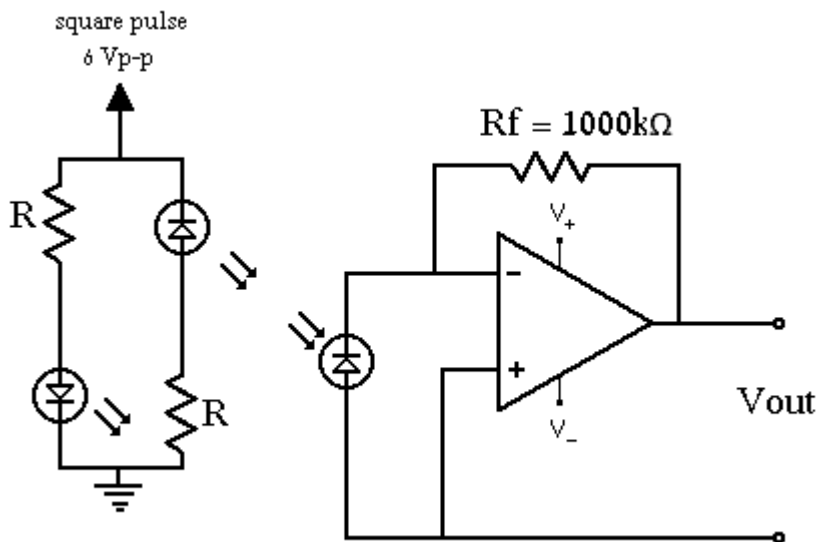
Figure 7: current to voltage converter of the sensor [10]

Figure 7 shows the current to voltage converter mechanism of the sensor. The photodetector in the circuit provides current as output where the current value reflects the intensity of the light coming out of the finger. The current passes through the resistor Rf and Vout = current x Rf.

The resistor Rf converts the current to voltage. As the output current is in the µA range, therefore a high resistor is being used to get a significant output voltage.

3.4 Hardware Design

The expected output of the sensor is in a function with frequency same as heart rate. Also, the output is likely to contain high frequency noise from the motion artifact and ambient light. Therefore a filtering system is required that will pass approximately 0 to 3 Hz frequency signal. To avoid eliminating necessary components from the signal, a low pass filter is designed for this problem that will pass DC to 10Hz. Appendix B discusses the design procedure in detail. Moreover, in the hardware design the filtered signal is passed through an amplifier that has gain of 10 approximately.
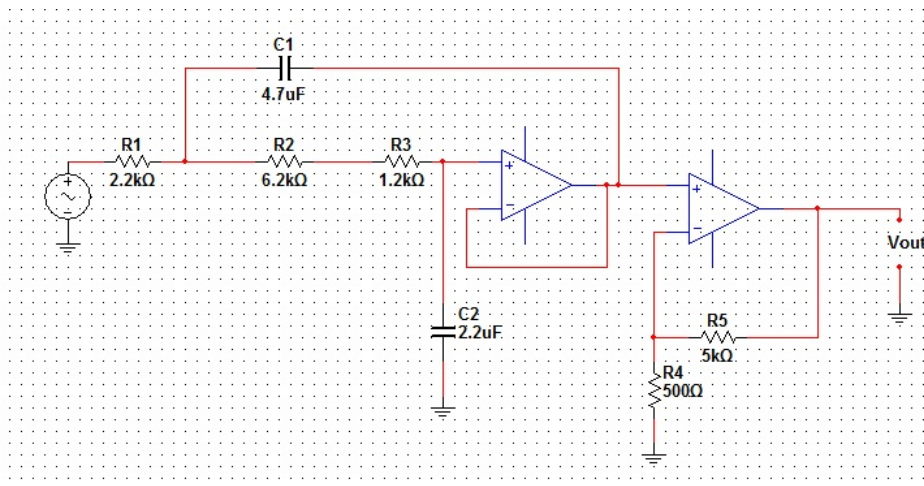
Figure 8: Final design of the circuit (generated using Multisim); C1 and C2 are capacitors of values 4.7µF and 2.2µF respectively and R1, R2, R3,R4 and R5 are resistors of values 2.2kΩ, 6.2 kΩ, 1.2 kΩ, 0.5 kΩ and 5 kΩ.

The designed filter was tested using Multisim software. Table 2 lists the values obtained by the simulation with 0.1 V peak-peak sine function.

Table 2: Output value of the designed circuit obtained using Multisim

Frequency (Hz)	Vin (V)	Vout(V)
2	0.1	1.090
4	0.1	1.060
5	0.1	1.030
6	0.1	0.999
7	0.1	0.958
8	0.1	0.911
9	0.1	0.858
10	0.1	0.808
11	0.1	0.746
12	0.1	0.690
13	0.1	0.635
14	0.1	0.584
15	0.1	0.536
20	0.1	0.352
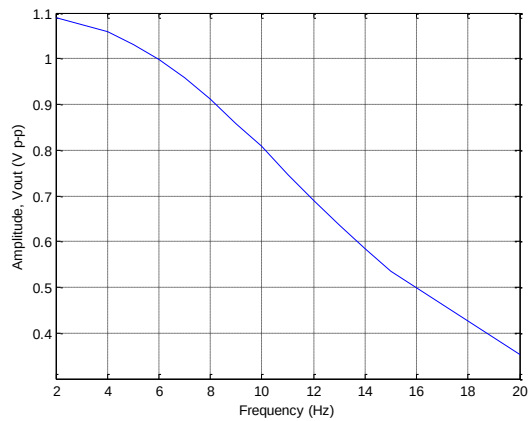
Figure 9: Plot of amplitudes from the Multisim data

Figure 9 is obtained by plotting the data from table 3 using MATLAB. It can be concluded that the design shown in figure 8 gives a low pass filter with cut off frequency approximately at 10 Hz as well as an amplifier with approximately 10 gain.

3.5 Software Design

Blood Oxygen Saturation Calculation

In this case, assume that the input of the data is digital form of the output of the designed hardware. Also, one LED cycle refers to a period of time where each (red and infra red) LED has been 'ON' once.
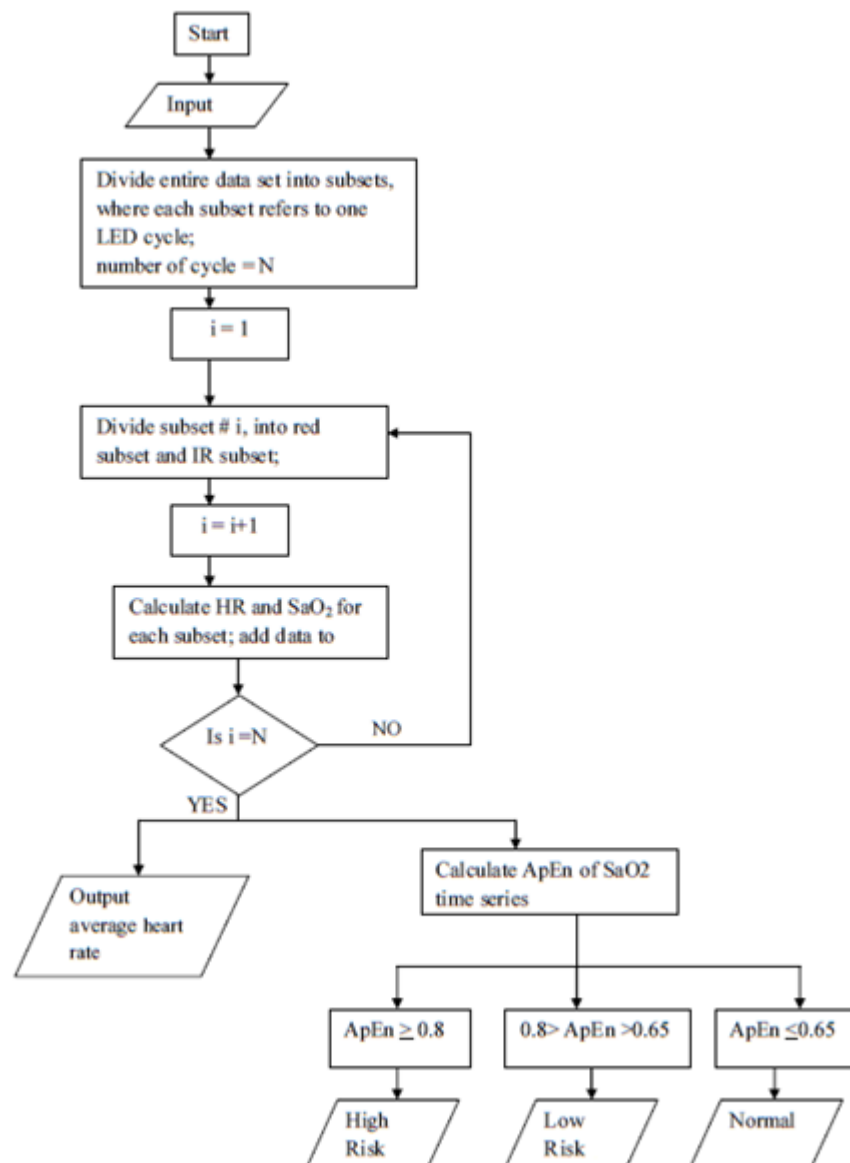
Figure 10: flow chart of the SaO2 calculator

Figure 2 shows the ApEn threshold for sleep apnea to be approximately 0.77, but in the case of this software the high risk threshold has been rounded off to 0.8 and the low risk threshold was adjusted to be 0.65.

4. Experimental Procedure and Development

The designed explained in section 3 was developed for this project. The hardware development was carried out in a laboratory where there was constant access to a oscilloscope and function generator as well as a +/-5V power supply.

4.1 Circuit Design and Testing

The designed circuit was developed in the lab using 5% resistors and OP491 op-amp [14]. Figure 28 shows the picture of the circuit developed on a breadboard. Table 3 shows the behaviour of the circuit. Vin represents the peak to peak amplitude of the input sine wave and Vout represent the same for the output sine wave. The test was performed in the lab using a function generator to generate the input sine wave and change the frequency to see the output behaviour.

Table 3: Output value of the designed circuit obtained using Multisim

| Frequency (Hz)	Vin (V)	Vout(V)
2	0.1	1.08
4	0.1	1.12
5	0.1	1.04
6	0.1	1.04
7	0.1	1.00
8	0.1	0.96
9	0.1	0.83
10	0.1	0.80
11	0.1	0.76
12	0.1	0.69
13	0.1	0.64
14	0.1	0.56
15	0.1	0.52
20	0.1	0.36
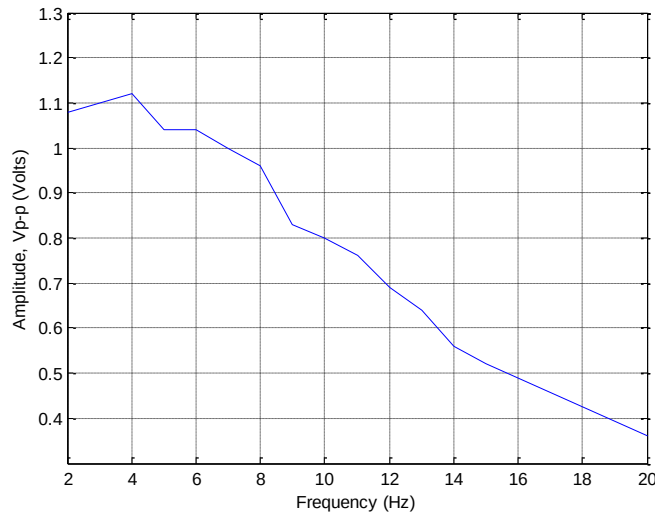
Figure 11: Plot of amplitudes obtained at the laboratory

Figure 11 shows a plot of peak-peak amplitude vs. frequency for data acquired in the lab. By comparing this plot with the ideal plot in figure 9, it was decided that significant similarities exist between the two, there. This also shows that approximately half of the amplitude is diminished when frequency is 10 Hz. Therefore it was decided that this circuit is appropriate for the development procedure.

4.2 Sensor Development and Hardware Testing

The already developed model to calculate SaO_2 assumes the wavelength of red light to be 640 nm and the wavelength of the infrared light to be 960 nm. However, due to availability, the used values were slightly different. The selected Red LED produced bright red light with 631 nm wavelength. The infrared LED produced infrared light of 940nm. The assumption in this case was that the small change in the wavelength of the LEDs will not make a significant difference in the SaO_2 calculation procedure.

Table 4: Parts used for the sensor

Part #	Part Description
RL5-R5015	Red LED, 631 nm [11]
SIR-320ST3F	Infra red LED, 940 nm [12]
BPW24R	Silicon PIN Photodetector , 600 – 1050 nm sprectral bandwidth

Initially the sensor was designed so that the LEDs and the photodetector are directly attached to the breadboard and the finger is placed in between. Figure 25 in appendix C shows the initial set up of the sensor. However, this method caused a large noise and the output contained spikes and missing data due to the movement of the finger and LED components. Moreover, the design was uncomfortable, hampered practicality to a large extent and did not provide stability for the development process. Therefore, the design was changed to a finger clip that will hold the photodetector and the LEDs tightly against the skin.

The LED in this circuit was powered using a function generator with square wave of 6V peak-peak amplitude and 100mHz frequency. Therefore each cycle with the each LED being 'ON' once was 10 seconds.

Figure 29 and 30 in appendix C shows the output of the hardware on a oscilloscope obtained during the development procedure.

4.3 Data Collection and software development

For the software requirement for this project, the data needed was collected using LabVIEW. The LabVIEW input probe was connected directly to the output of the signal amplifier. The sampling frequency was set to 1000 samples/s. The data collected was in the form of a digital time series and was used directly by the designed software.

The data collected by LabVIEW was saved in a text file. This file was used as the input for the software. The manipulation steps are described below. The following description is for execution of the software. The software matches the flowchart in figure 10.

4.3.1 Division of Data

The data collected contained several square wave pulses depending on the total test time. The higher part of the square wave contained information regarding the infrared data and

the lower portion contained the red data. The data collection in LabVIEW did not start at the beginning of a square wave pulse. Therefore, for calculating SaO_2 only the complete pulse sequences were selected starting with the red light sequence. The incomplete cycle at the end was also eliminated at the revision process.

Figure 12 contains data for a test that was approximately 140 seconds long. Figure 13 shows data that had been revised and selected for the calculation. The signal appearing in figure 13 did not contain any incomplete cycle. The amount of neglected data was less than 20 seconds in duration regardless of the test constraint as only two incomplete cycles were eliminated. Note that one cycle is 10 seconds long.
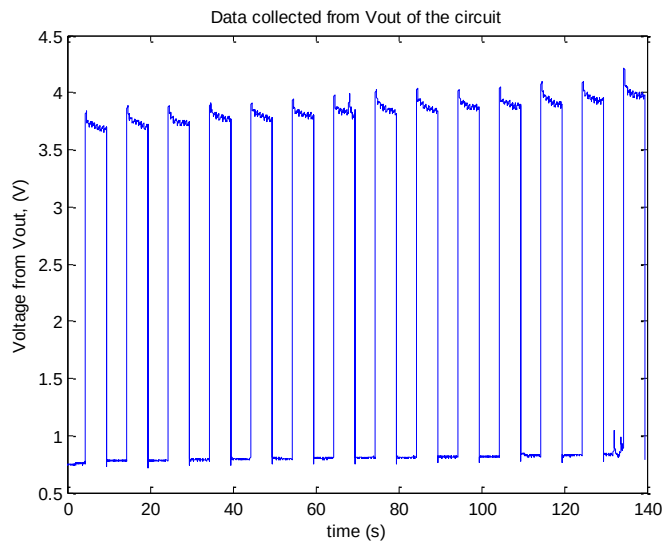

Figure 12: Actual Data (Collected from the output of the hardware circuit)
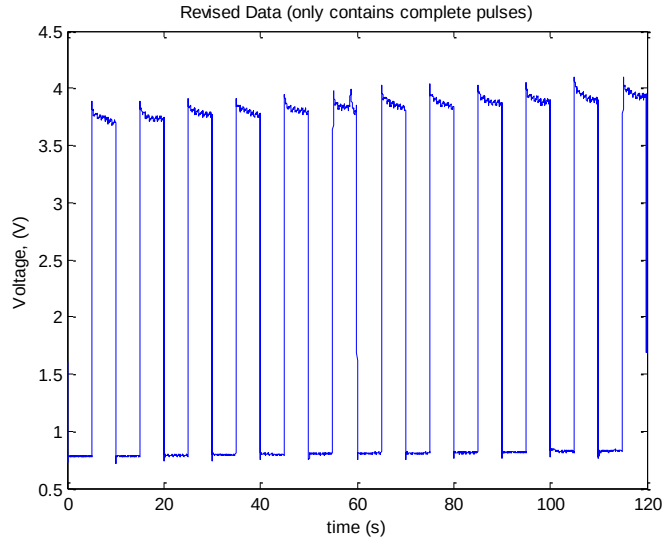
Figure 13: Revised Data (Only containing complete LED pulses)

After the useful data was selected, every pulse sequence was divided into red and infrared subsets. Before applying equation (2) to the dataset, the infrared and the red signals were normalized. As the DC offset of the IR signal was higher, the IR signal was the only one normalized to the level of the red signal.

$$IRnorm = IR - (meanIR - meanR) - - - -(5)$$

Where,

      IRnorm = normalized IR signal

      IR = Collected infrared signal

      meanIR = mean value of the IR signal

      meanR = mean value of the Red signal

Equation 5 was used in the normalization process. After the signal was normalized, the peaks were detected and stored for further calculation.
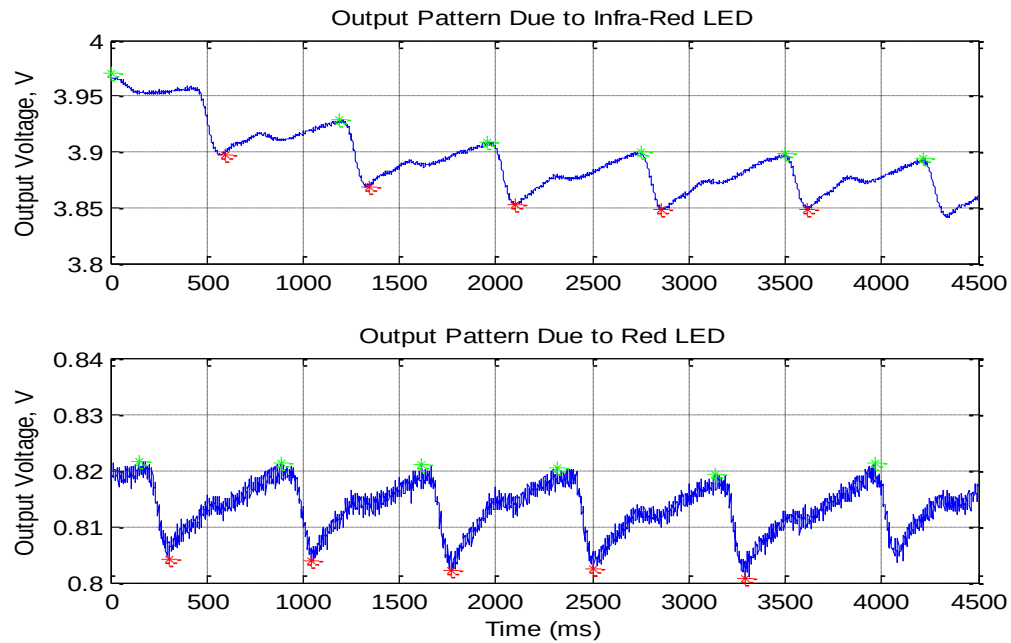
Figure 14: Red and infrared data separated (not normalized)
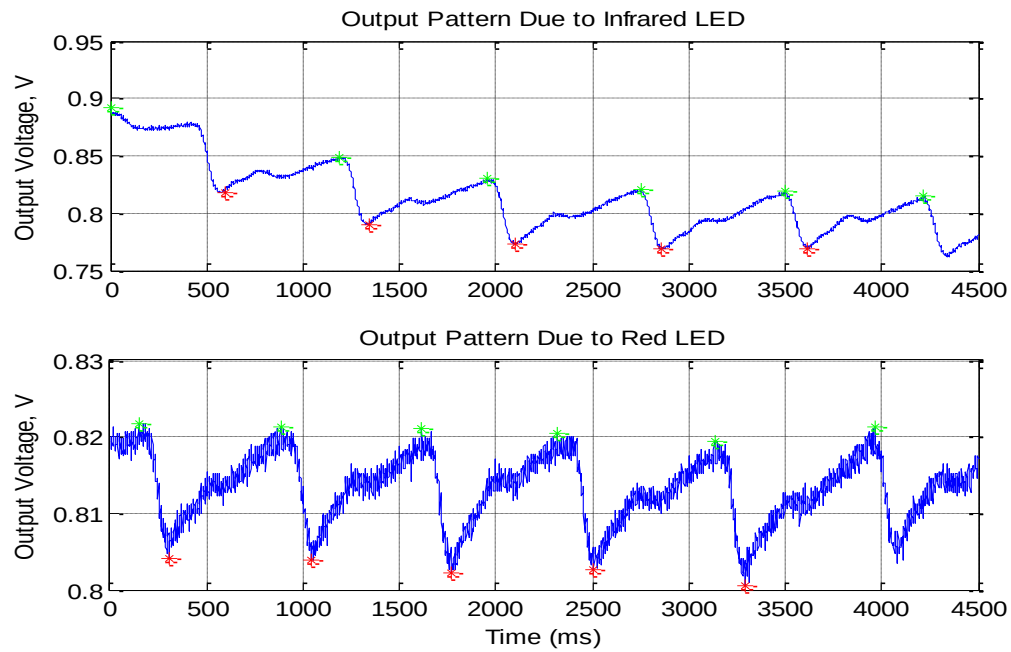
Figure 15: Red and infrared data separated and normalized

Figure 14 shows the data before normalization and Figure 15 shows the data after normalization. The red and green dots shown are the markers of the detected peaks. These

peaks provided information regarding the heart rate calculation as well as the variables for equation 2.

4.3.2 Calculating Heart Rate and SaO$_2$

The peaks from the previous step were used to calculate SaO$_2$ by using equation (2) and (3). Heart rate was calculated using equation (4). The software was designed so that the heart rate was averaged for the entire dataset and outputted as a mean result for the total study period. The SaO$_2$ was outputted as a time series for every pulse and approximate entropy analysis was done on the stated time series.

4.3.3 Approximate Entropy Calculator

The calculator developed to calculate ApEn followed the algorithm described in appendix A. A pre-solved example was used to test the validity of the designed software. The time series that was used for the testing purpose was a periodic time series for which the duration of each period was 5 seconds. For one period, the time series was x(t) = [61, 62, 63, 64, 65] and the duration of the total series was 50 seconds. The approximate entropy of the time series for m = 5 and r = 2 was given as 0.00189 [9]. The designed MATLAB software was used to calculate the approximate entropy of the same time series where the result was 0.0019, thus proving the computational validity.

4.4 Challenges During Experiment and Sources of Error

Several challenges were faced during the experiment and development procedure. Many of the data sets contained peaks that did not match the expected waveform. Experiments show that these types of peaks appeared during excessive motion of the finger.
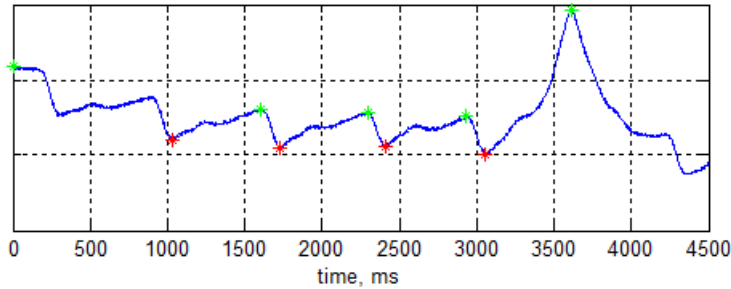
Figure 16: Example of a signal with motion artifact (undesired peak at approximately t = 3600)

Figure 16 shows an example of a peak acquired while data collection. Due to the sensor being attached to the breadboard by wires, the connection was not very sturdy and sometimes the motion atrifacts were difficult to control and produced a source of error in the final result. Data with error similar to figure 16 was inputted in the software and the software assumed the data to be ideal, but the produced result included errors.

Also, the designed sensor did not work for breath-holds. If during the test situation in the lab, breathing was paused then the infrared signal reduced as expected, however, the red signal became very noisy, where the noise amplitude was larger than the expected amplitude. This problem occurred for every trial to calculate SaO_2 for a sleep apnea simulation. Figure 17 shows a set of data collected while similar trial procedure.
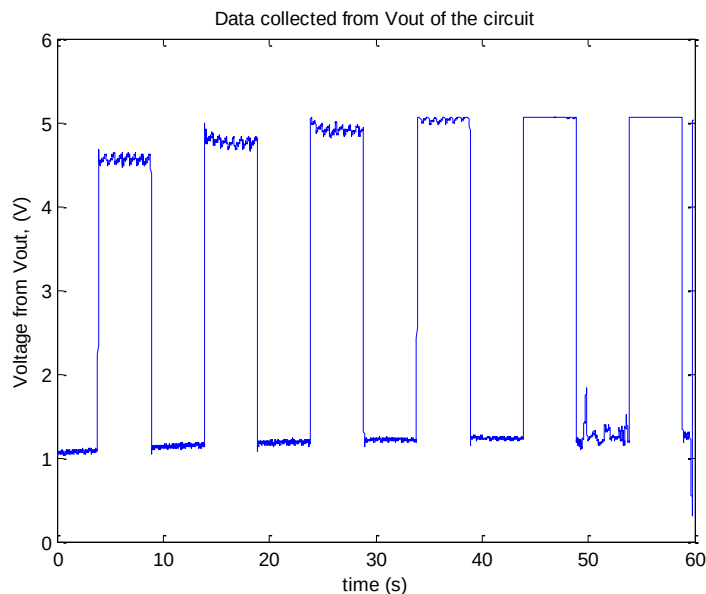

Figure 17: Sleep Apnea Simulation Trial Data

5. Results

The result of the developed device can be divided into three sections. The first set is series of SaO2 for the entire test procedure. This result is not important for the patient's viewing therefore the software does not generate a noticeable output containing SaO2 information. However, this series might be important to a medical professional to assess the patient's situation. The second set of result is the approximate entropy and the risk warning for a certain dataset. The risk warning is the most important result for a patient without a medical background. The third result is the heart rate for overnight study.

5.1 SaO$_2$

By using the data from figure 13, the following SaO$_2$ series was found. Note that the series contains 12 elements. This is due to the fact that the useful signal (Figure 13) duration was 120 seconds with each pulse being 10 seconds long. Each SaO$_2$ value corresponds to one cycle of red and infrared light.

Table 5: SaO2 time series example (Data collected from the output of the designed software)

Cycle #	SaO$_2$ (%)
1	99.6092
2	99.8450
3	97.6656
4	97.9422
5	97.5168
6	98.8377
7	98.3776
8	97.7214
9	98.5778
10	99.2596
11	98.3312
12	98.6281

For a normal human being, the blood oxygen saturation is 90% - 100% therefore the collected data was considered correct.

5.2 Sleep Apnea Detector

Due to the limitations in the sensor, the data could not be collected for a long period of time. For the test dataset, the data was collected for a little over than 2 minutes. This situation did not depict the actual test situation because a sleep study needs to be conducted for longer. Therefore to test the sleep apnea detection methodology, the data was repeated 100 times therefore the SaO_2 in the sleep apnea test was for 12000 seconds or 200 minutes. Figure 18 shows the SaO_2 time series that was the same as table 5 but repeated and was used in the ApEn calculator. The approximate entropy of the data was calculated to be 0.40546. As discussed earlier, sleep apnea is noted to be positive for approximate entropy higher than 0.8, therefore this case shows "normal" as a result. The output of the designed device includes two windows shown in figure 19 and figure 20 showing the summary of the sleep study.
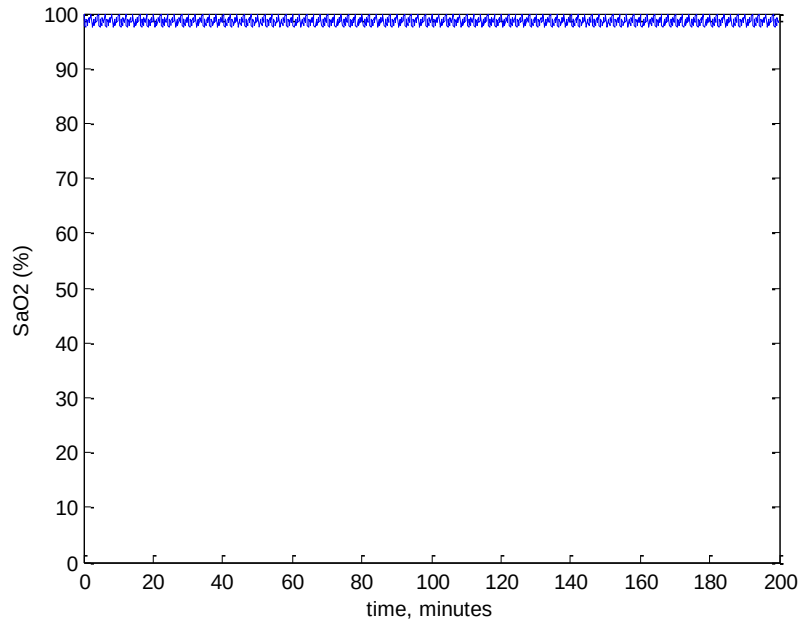

Figure 18: SaO2 time series for a healthy case scenario
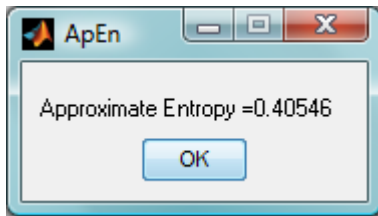
Figure 19: ApEn result for a healthy case scenario
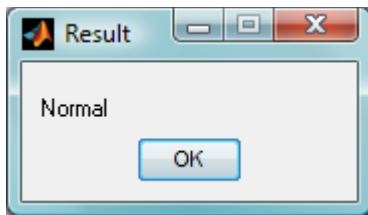
Figure 20: Overall condition of the healthy case scenario

Figure 1 (b) shows the pattern of SaO_2 fluctuation for a sleep apnea positive scenario. Similar dips in the SaO_2 time series were simulated to test the approximate entropy calculator. Also it can be observed in figure 1 that the overall SaO_2 data (other than the dips during apnea attack) is a little lower and noisier in the sleep apnea case therefore an overall reduction was introduced in the data where the reduction parameter varied randomly from 1%- 6% thus creating an extra noise component.
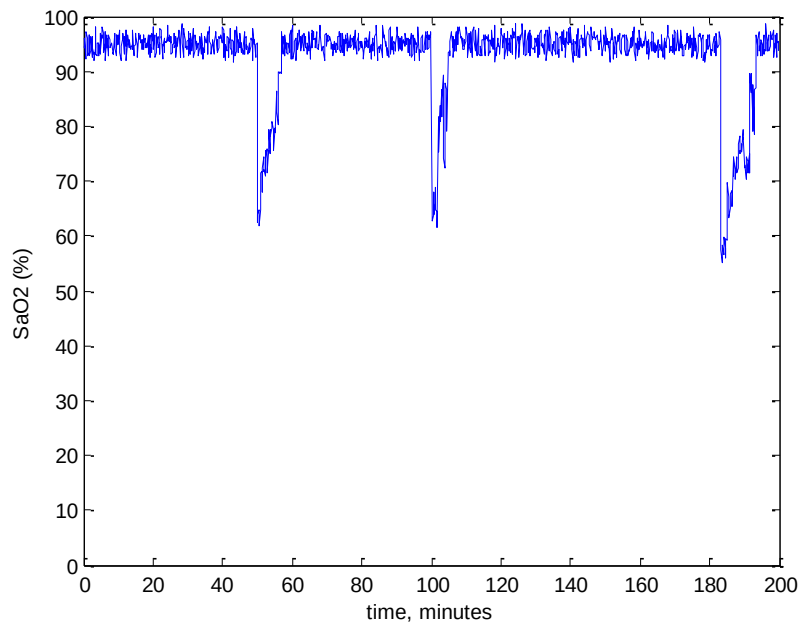
Figure 21: Noisy SaO$_2$ dataset simulating a sleep apnea case scenario
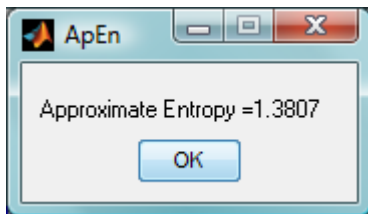
Figure 22: ApEn result for an apnea case scenario
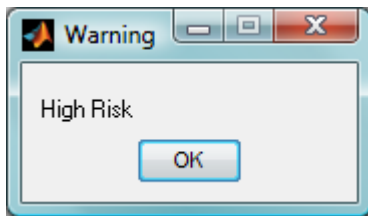
Figure 23: Warning window for an apnea case scenario

Figure 21 shows a data set that was used to verify the designed calculator. This dataset included the noise components and the dips discussed earlier. It can be interpreted as a patient having apnea attack after approximately 50, 100 and 180 minutes after the test began. Figure 22 shows the ApEn value for this case to be 1.3807 which is as expected

very high as the simulated apnea attacks are very close to each other. For this type of results the software generates a 'High Risk' warning (figure 23).

5.3 Average Heart Rate

The designed device gives an average heart rate for the entire study period. The following window is an example of the heart rate output window.
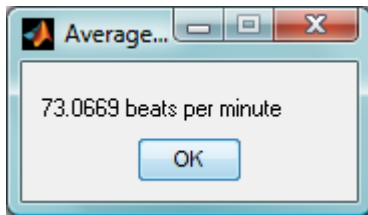
Figure 24: Average heart rate window

6. Future Development

Several upgrades can be made in the obtained device that will increase efficiency. If all the upgrades are properly installed then a few features can be added to improve the usability of the device. These features and upgrades are discussed in the following subsections

6.1 Possible Upgrades

For the scope of this project, the red and infrared LEDs of the sensor were powered using a function generator. This practise caused addition of one extra equipment and hampered practicality. Use of a 555 timer chip to generate a similar square wave can eliminate the use of the function generator and make this device more practical. [15]

LabVIEW was used to transfer data from the output of the circuit to the software. This procedure is not efficient as the data transfer is happening in real time and a wire connection is introduced from the test subject to a computer adding constraint and discomfort. Moreover, LabVIEW is not a common software and for this project, it is only practical to be used in the development procedure. The data transfer can be done using wireless or Bluetooth which will reduce the discomfort factor. Nowadays, Bluetooth and wireless receivers are common in personal computers and cellular phones. This upgrade might create an easy solution but introduce additional expenses, higher processing power and the requirement of a special receiver. A more economic solution of this problem is attaching a memory card to the hardware design that will store the data for further transfer to a suitable calculating module such as a laptop, PDA or cellular phone.

To reduce cost of this project, a cloth's pin was used as the base of the finger clip sensor. For the scope of the project, this arrangement was sufficient however improvements in the finger clip must be made to obtain better results. The two major problems with this design were ambient light interference and motion artifact. The ambient light effect can be reduced by adding a cover to the finger clip design. The motion factor was mainly created by the movements of the wires in the breadboard and movement of LEDs due to

abrupt motion of the finger. The wire factor can be eliminated by attaching the wires to a printed circuit board. The finger motion factor is more serious as this type of movement is common during sleep. This problem can be reduced by either using a finger clip that holds the finger from all sides or by attaching the sensor system to a glove that will be worn during the sleep study.

6.2 Additional Features

Given that the previously mentioned upgrades and solutions are applied, the following features could be added to the device to make it more efficient.

The designed device can be used for monitoring purpose therefore adding a real time display monitor can prove to be beneficial. Also, people with higher probability of sleep apnea might want to communicate their results with medical professional. A system can be developed that will email the overnight study result to the management facility of the medical records.

7. Conclusion

The device developed in this project is a successful prototype that correctly predicts sleep apnea risk. However, it must be noted that this prototype device is not a replacement for the current detection methods for sleep apnea. More research needs to be conducted to get stronger connection between SaO_2 and AHI to obtain a more accurate testing system for OSA. This device can also be used for a few other purpose other than sleep apnea sleep study. It can also be used for monitoring patients with breathing disorder as well as other illness that might affect SaO_2. It can also include features that will allow it to measure the progress in sleep apnea treatment.

The primary goal of the project is successfully accomplished but there were assumptions that were used to bypass a few presented problems. The pulse oximeter designed only worked for a healthy human being and could not produce results for a sleep apnea case. The output of this project is very similar to the expectations. The major shortcomings were mostly caused by simplification in the oximeter design. Given enough resources to overcome the aforementioned issues, this prototype can be practically used for both medical and research needs.

Overall, this project provided deeper understanding of various biomedical engineering concepts such as sensor design, instrumentation procedure as well as challenges faced as an engineer. It definitely created a profound appreciation for the field and understanding of a complete project construction.

Appendix A

Approximate Entropy [5], [9]

Approximate entropy, ApEn is a statistical tool that quantifies the fluctuation unpredictability of a time series. Higher ApEn value of a time series means higher irregularity in the data.

Two important parameters in calculation ApEn of a time series are, tolerance window, 'r' and run length or pattern length 'm'. To summarize, ApEn measures the possibility of a run patterns to close within the same tolerance window 'r'.

In the following steps, the calculation of ApEn will be developed assuming a time series x that has N elements.

Step1: The time series x is divided into N-m+1 subseries using the following equation

$$X(i) = [x(i), x(i + 1), ... , x(i + m - 1),$$

Where, i =1,2,.....N-m+1 and X represents the constructed subseries

Step 2: A matrix N_m of size [(N-m+1) x 1] is formed where N(i) represents number of existing subseries X that are similar to X(i). In this case, the parameter 'r' is used to determine similarity. Two of the subseries is considered similar, if the difference between each element of the two subseries is less than or equal to r.

Step 3: Another matrix C is formed where,

$$C = \frac{N_m}{(N - m + 1)}$$

Step 4: A parameter Φ that represents the mean of the values of the matrix C.

Step 5: Step 1 to 4 is repeated once for m = m+1

Step 6: Approximate entropy is calculated using the following equation

$$ApEn = \frac{\Phi_m}{\Phi_{m+1}}$$

Appendix B

Design of the LPF and Amplifier [16], [17], [18]

Low Pass filter

The filter used in this case was a 2nd order low pass Bessel filter using Sallen-Key topology with 10 Hz cut-off frequency.
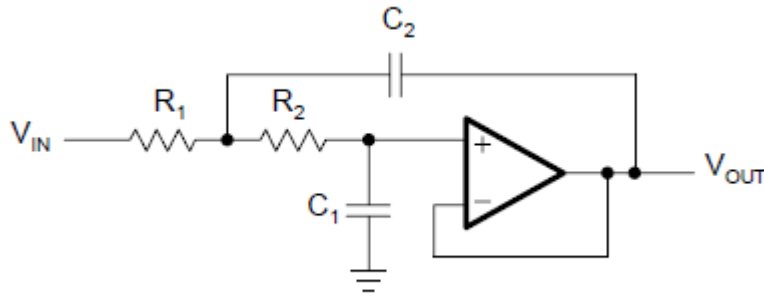

Figure 31: Unity gain low pass filter [17]

The condition of selecting C_1 and C_2 is,

$$C_2 \geq \frac{C_1 x4b}{a^2} ----- (6)$$

If C_1 and C_2 are preselected then the resistors can be calculated using,

$$R_{1,2} = \frac{aC_2 \mp \sqrt{(a^2 C_2^2) - (4bC_1 C_2)}}{4\pi f_c C_1 C_2} ----- (7)$$

In this case,

a = 1.3617

b = 0.6180

$f_c = 10$ Hz

If C_1 and C_2 are selected to be 2.2µF and 4.7µF respectively, the condition given by equation 6 is satisfied. Using equation 7 and the capacitor values, R1 = 1.94kΩ, R2 = 7.9kΩ

However considering availability, the selected resistor values were, R1 = 2.2kΩ, R2 = 7.4kΩ

Non-inverting amplifier

The following non inverting amplifier design was used for the hardware
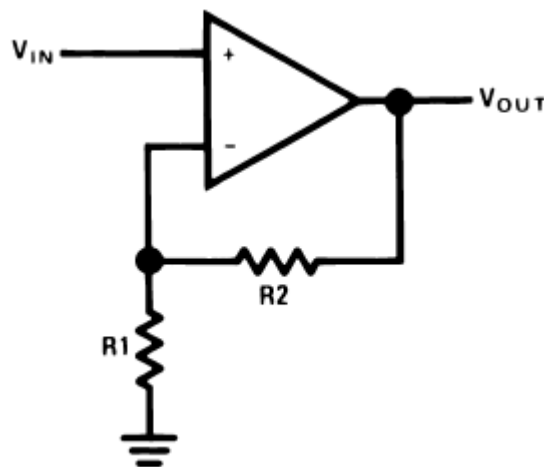
Figure 32: Non inverting amplifier

Selected gain = 10

The following equation is used to calculate gain of the amplifier.

$$Gain = 1 + \frac{R_2}{R_1}$$

R_2 was selected as 5kΩ

Therefore,

$$R_1 = \frac{R_2}{G - 1}$$

R₁ = 555.56Ω

Due to availability the R_1 for the non inverting amplifier is selected to be 500Ω which changed the gain to 11. This minor mismatch of result does not affect the output.

Appendix C

Hardware Images

The following images were captured in the laboratory during the development procedure
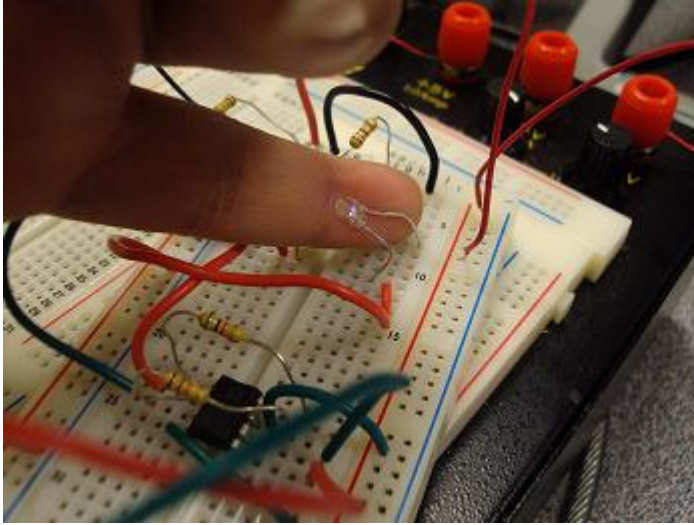

Figure 25: Initial set up of the sensor

The figure above shows how the sensor was set up at the beginning of the design process. The LED was on top of the finger and attached directly to the breadboard and the photodetector was on the opposite side also attached directly to the breadboard. The following figure (figure 26) shows the upgraded design of the test sensor. The red and infrared LED was attached to one side of a cloth's pin and the photodetector was attached on the other side. Also long wires were soldered to the ends of the photodetector and the LEDs providing a bigger motion space for the finger during a test procedure. Figure 27 shows the how the finger should be connected to the sensor.
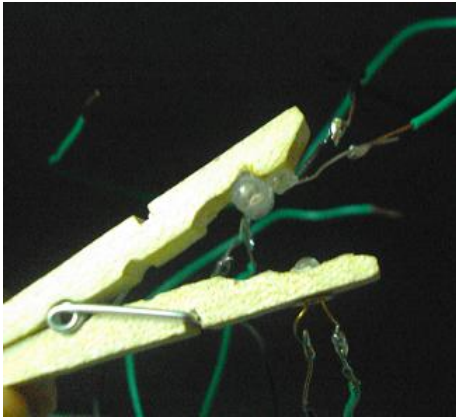
Figure 26: Picture of the sensor
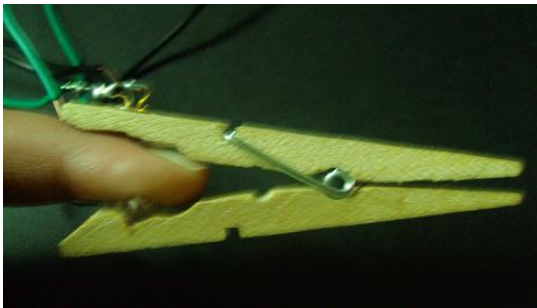
Figure 27: Picture of the sensor in use
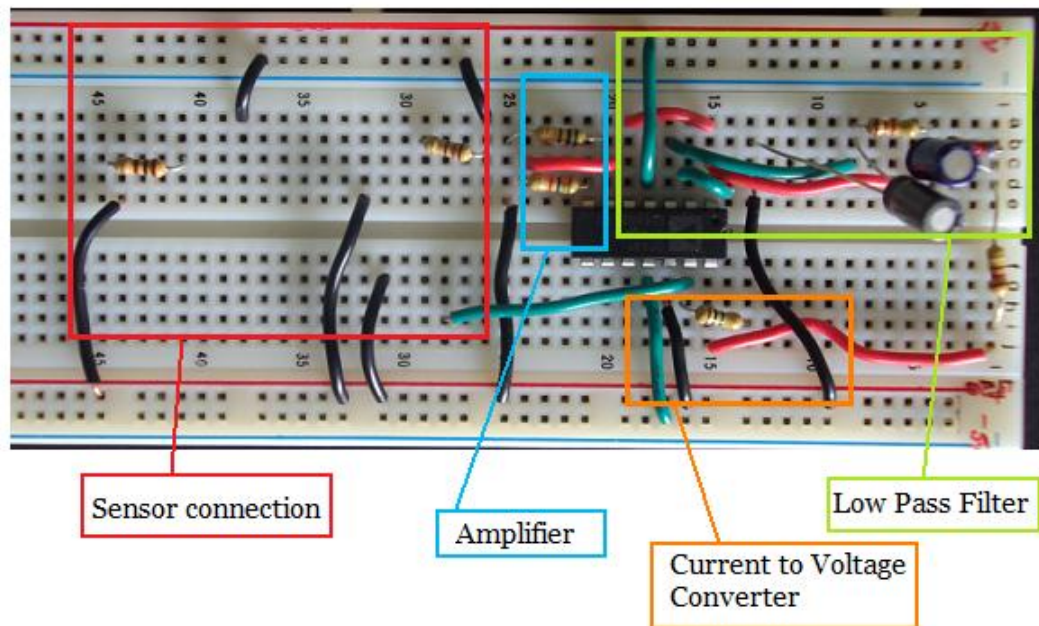
Figure 28: Picture of circuit

Figure 29 and figure 30 shows the output observed in the lab using an oscilloscope during the development procedure.
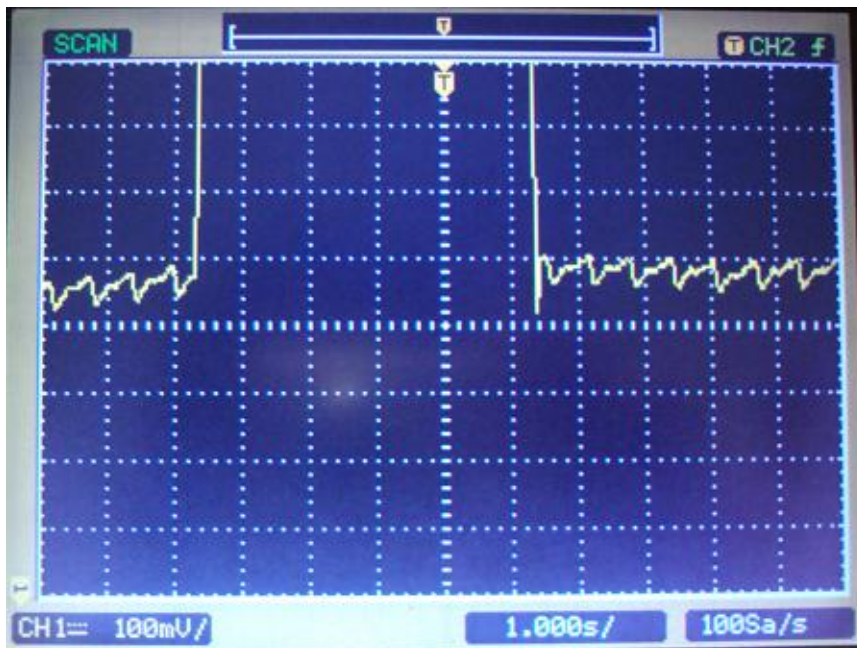

Figure 29: Red light output in the oscilloscope
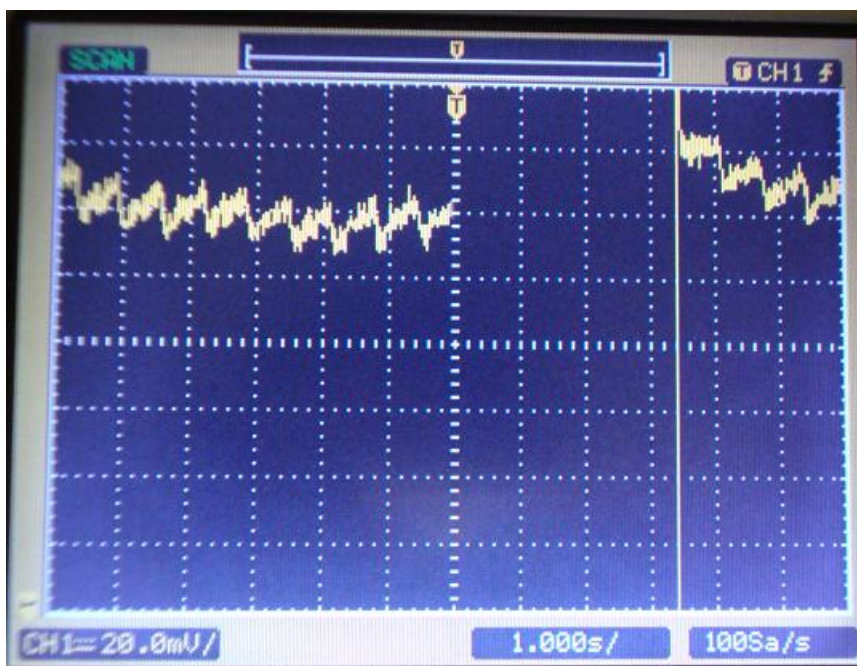
Figure 30: Infrared output in the oscilloscope

Appendix D

Parts used in the device development procedure

Table 6: Completer parts list

Part	Price ($)	Available at
Red Led, RL5-R5015	0.24	www.superbrightleds.com
Infrared LED, SIR-320ST3F	1.06	www.digikey.com
Photodetector, BPW24R	3.83	www.digikey.com
OP491	6.98	www.digikey.com
Resistors	5.00 (approx.)	Assorted pack
Capacitors	3.00 (approx.)	Nutech Electronics
Function Generator	N/A	Course Laboratory
Oscilloscope	N/A	Course Laboratory
Breadboard	N/A	Course Laboratory
LabVIEW data aquisation system	N/A	Course Laboratory

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