VOL. 10, NO. 17, SEPTEMBER 2015 ISSN 1819-6608
ARPN Journal of Engineering and Applied Sciences


www.arpnjournals.com

INTEGRATION OF LOW COST SpO_2 SENSOR IN A WEARABLE MONITOR

Ajith K. G.[1], Bony George[1], Aravind B.[2] and Martin K. M.[1]
[1]NIELIT, Calicut, Kerala, India
[2]Mobilexion Technologies, India
E-Mail: ajithkallidukkil@gmail.com

ABSTRACT

Pulse oximetry is the non-invasive measurement of the oxygen saturation (SpO_2). It is used for a rapid assessment of a patient's respiratory function to determine onset of hypoxemia (oxygen starvation) or COPD (Chronic obstructive pulmonary disease). The sensing device is worn in the finger-tip which makes it the one with least discomfort among the devices for measuring vital parameters. In many critical situations, it is the ideal candidate for continuous monitoring using a wearable system for alert generation. We have developed the prototype of such a device which can be connected to any android based mobile phone. It consists of a hardware subsystem with a very small footprint for continuous acquisition of SpO_2 signal for transfer to the mobile phone of the user. A part of the mobile application runs in the background to continuously acquire this signal, convert it into calibrated values of SpO_2 and pulse rate and store them in the local DBMS. In case the connection to a Hospital DBMS is enabled, it is updated there as well. The foreground part of the application provides the instantaneous values of pulse rate and SpO_2, along with their trends and minimum values. By continuous monitoring of pulse rate and SpO_2 in the background, the application can detect the onset of hypoxemia and COPD and can give suitable alarms. The Hospital DBMS can be accessed by the consulting clinicians, thereby allowing remote monitoring of patient health condition. The device is being integrated into a wearable Body Area Network having ECG, Pressure and Temperature sensors for the complete monitoring of all vital parameters. Once clinically accepted, it can become a low cost alternative to the current bedside monitoring system used in hospitals. This paper presents the hardware and software aspects of the SpO_2 sensor segment developed.

Keywords: SpO_2, body area network, pulse oximetry, hypoxemia, COPD.

INTRODUCTION

One of the most important elements needed to sustain life is oxygen (O_2) because it is used by cells to turn sugars into useable energy. Oxyhemoglobin (HbO_2) is the protein hemoglobin, found in red blood cells, bounded to O_2 that delivers 98% of oxygen to cells. The measurement and calculation of the percentage of HbO_2 in arterial blood is known as oxygen saturation (SpO_2).

Originally, SpO_2 was measured by taking samples of blood and measuring O_2 levels directly. This method was invasive and was unable to provide real-time measurements. Due to this, SpO_2 was not recognized as an important measure of wellness until a non-invasive method of measuring it in real-time (pulse oximetry) was established.

The first device used to continuously measure blood oxygen saturation of human blood in vivo (SaO_2) was built by Karl Matthes in 1935 [1]. In 1983, William New and Mark Yelderman, produced the pulse oximeter with the aim of making it an intra-operative monitoring device [2].

Pulse oximeter uses a light emitter with red and infrared LEDs that shines through a reasonably translucent site with good blood flow. Absorption due to tissue, skin or muscle remains fairly constant, whereas absorption due to arterial blood varies. Arteries expand due to the pumping of the heart, thereby increasing the tissue between the LEDs and the photo-diode and the light absorption. The blood-stream is affected by the concentration of HbO_2 and Hb, and their absorption coefficients are measured using two wavelengths 660 nm (red light spectra) and 940 nm (infrared light spectra).

Pulse oximetry derives SpO_2 and pulse rate (PR) from photoplethysmogram (PPG) signal that reflects the change in vascular blood volume with each cardiac beat. It is obtained by measuring changes in light absorbed by the blood. Red and infrared wavelengths are used to obtain the PPG because these wavelengths are easily transmitted through tissues.

Deoxygenated and oxygenated hemoglobin absorb different wavelengths, with former having a higher absorption at 660 nm and latter having a higher absorption at 940 nm. Pulse oximeter uses this difference to deduct the SpO_2 value and the pulse rate. It allows accurate determination of O_2 levels in patients that are sedated, anesthetized, unconscious, and unable to regulate their own oxygen supply. It provides vital information needed to avoid irreversible tissue damage.

Once a patient starts losing oxygen, a doctor has less than three minutes to prevent risk of brain damage, heart failure and death. A healthy body should never fall below 95% oxygen saturation, and oximeters can detect changes as small as 1%. Before oximeters were invented, experienced clinicians would only notice signs of hypoxia when patients became cyanosed – literally their skin begins turning blue, with oxygen saturation damagingly low at 85%. It is no wonder that Pulse oximetry became a standard procedure for the measurement of blood oxygen saturation in hospitals and SpO_2 got accepted as the fifth

vital sign (in addition to temperature, blood pressure, pulse, and respiratory rate) in clinical assessment.

Our interest in SpO_2 originated from an attempt to develop a low cost bed side monitoring system that measures the patient vitals, continuously, in the hospital environment. Even though, the bed side monitor is an invaluable diagnostic tool, their widespread usage is hampered due to the high cost and the requirement of wired connections to the body sensors that impose movement restrictions.

We are in the process of developing a wearable monitoring system (e-PMS) that would solve both the above. It consists of a vest that houses the various measuring elements routed to an Audrino based central system connected to a smart phone kept in a specially designed pocket in the vest. Details of the first version of this are given in [3]. In our new version, the Tablet PC meant as the analysis platform is replaced by the smart phone, which reduces the cost even further.

One of the important components of e-PMS is the SpO_2 and pulse rate measurement segment that has been developed from low cost components, as part of this work. This paper presents the details of this. It is organized as follows. Section II describes the theoretical basis of obtaining the PPG signal and the measurement of SpO_2 and PR from it. Section III describes the system hardware developed. Section VI gives a brief overview of the system software for data acquisition, storage and alert generation. Section V describes the display software for the smart phone. Finally, section VI gives an overview of the design of the vest.

THEORETICAL BASIS

Figure-1 gives the theoretical basis for pulse oximetry. A light emitter with red and infrared LEDs shines through a reasonably translucent site with good blood flow (finger, toe, ear lobe, etc.).

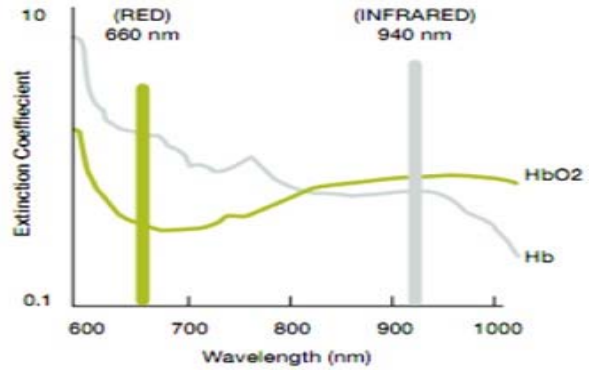

Figure-1. Blood light absorption graph.

A photo-detector receives the light that passes through the measuring site. There are two methods for this - transmission and reflectance. In the transmission method (Figure-2), the emitter and photo-detector are opposite of each other with the measuring site in-between. In the reflectance method (Figure-3), the emitter and photo-detector is next to each other on top the measuring site.

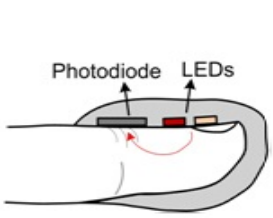

Figure-2. Reflection type.

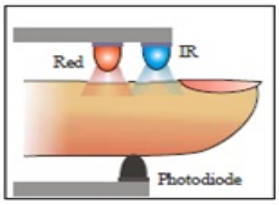

Figure-3. Transmission type.

The absorbance of light at a specific wavelength by a homogenous solution can be accurately determined by the Beer-Lambert's law:

$$I_t = I_0 \cdot e^{-\alpha c d}$$

Where I_t is the transmitted light intensity, I_0 is the incident light intensity, α is the specific absorption coefficient of the sample, c is the concentration of the sample, and d is the path length of light transmission.

In oximetry, it is assumed that blood is a two-component homogeneous mixture of Hb and HbO_2 and light absorbance of these components is additive. However, other variables in the biological media such as bone, skin, tissue, muscle and blood also scatter light. The absorption of light also depends on both skin thickness and color. Beer-Lambert's Law is unable to account for all of these variables.

Modern pulse oximetry relies on the detection of PPG signal produced by variations in the quantity of arterial blood associated with periodic contractions and relaxations of the heart. As shown in Figure-4, the magnitude of the PPG signal depends on the amount of blood ejected from the heart with each systolic cycle, the optical absorption of blood, absorption by skin and various tissue components, and the specific wavelengths used to illuminate the vascular tissue bed.

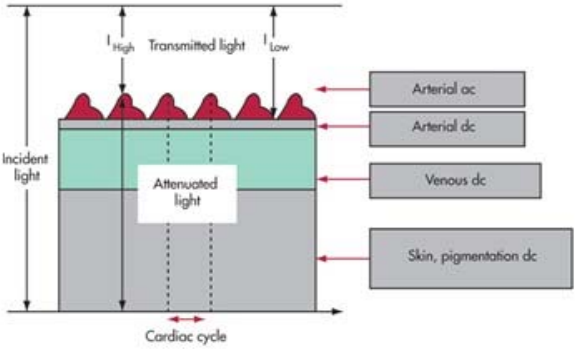

Figure-4. Variations in light attenuation by tissue.

During systole, when the arterial pulsation is at its peak, the volume of blood in tissue increases. This additional blood absorbs more light, thus reducing the light intensity which is either transmitted or backscattered.

During diastole, less blood is present in the vascular bed, thus increasing the amount of light transmitted or backscattered. The pulsatile part of the PPG signal is considered as the "AC" component, and the non- pulsatile part, resulting mainly from the venous blood, skin and tissue, is referred to as the "DC" component. A deviation in the LED brightness or detector sensitivity can change the intensity of the light detected by the sensor. This dependence on transmitted or backscattered light intensity can be compensated by using a normalization technique where the AC component is divided by the DC component. The time invariant absorbance due to venous blood or surrounding tissues does not have any effect on this ratio. This normalization is carried out for both the red (R) and the infrared (IR) wavelengths. It has been shown that, the normalized R/IR "ratio of ratios" as given in the equation below,

$$\frac{R}{IR} = \left(\frac{AC_R}{DC_R} \Big/ \frac{AC_{IR}}{DC_{IR}} \right)$$

is linearly related to SpO_2. Pulse oximeter measures absorbance at two different wavelengths and are calibrated from the data collected by looking up a value for SpO_2, using the empirical relationship given by the equation

$SpO_2 = A - B(R/IR)$

Where A and B are linear regression coefficients which are related to the specific absorptions coefficients of Hb and HbO_2. The constants and are derived empirically during in-vivo calibration by correlating the ratio calculated by the pulse oximeter against SpO_2 from arterial blood samples by an in vitro oximeter for a large group of subjects. Pulse oximeter reads the SpO_2 of the blood accurately enough for clinical use under normal circumstances because they use a calibration curve based on empirical data.

SYSTEM HARDWARE

We use transmition method of pulse oxymetry where the photo detector and LEDs (Red & IR) are placed at opposite sides. The current output from the photo detector is converted to a curresponding voltage value using a current to voltage converter. This is amplified, filtered and given to the microcontroller in an Arduino uno board. (The Arduino Uno is based on the ATmega328. It contains 32 KB of Flash memory, 2 KB of SRAM and 1 KB of EEPROM.It has 14 digital input/output pins (of which 6 can be used as PWM outputs), 6 analog inputs, a 16 MHz ceramic resonator, a USB connection, a power jack, an ICSP header, and a reset button). Microcontroller reads the analog value and convert this to curresponding digital value.

Figure-5(a) gives the pre-processing segment consisting of LED, voltage conversion, amplification and filtering with cutoff frequency of 2.3 Hz. The microcontroller (see Figure-5(b)) performs the Analogue to Digital Conversion, and short term local storage of the digital signal streams into packets that are transmitted to the smart phone over the micro USB communication line running through the vest. The smart-phone computes the R/IR ratio and consults the calibration curve to compute SpO_2. It calculates the pulse rate by counting the peaks in the PPG signal. These are stored in the local data base and are fed to a prediction engine to generate local alarms. This uses the smart phone's loud speaker to give warning indicators if the calculated SpO_2 or pulse rate falls below the values prescribed by clinician.

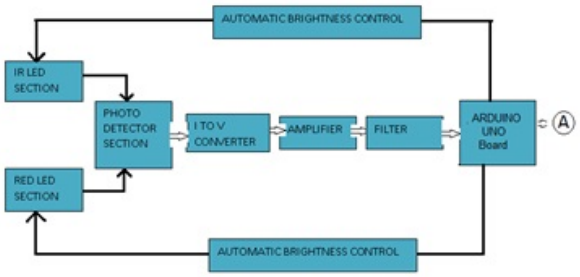

Figure-5(a). Block diagram of pre-processig section.

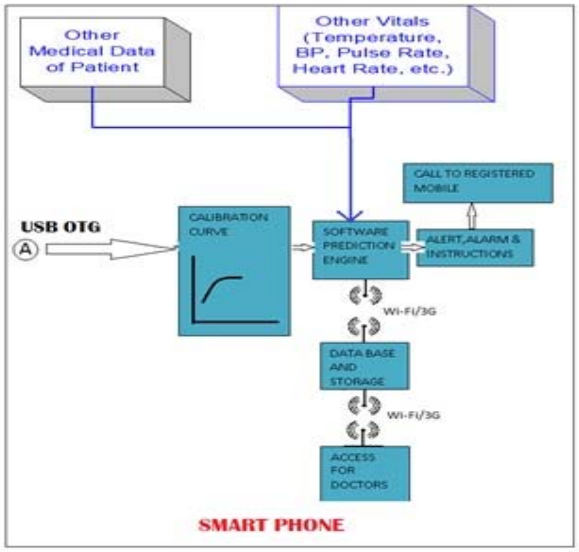

Figure-5(b). Block diagram of pre-processig section.

Figure-6 gives the LEDs driving circuit. In order to have an accurate control over the intensity of the LEDs we use two PWM waves for driving them by changing their duty cycle. An over current protection using additional transistors (Q5 and Q6) are provided since photo devices are very sensitive to the driving current. Figure-7 gives the trans-impidance amplifire and filter amplifier circuit. The former is used to convert the current output of the photo detector to the curresponding amplified voltage value. The output of the photodetector is in nanao-Amp range and amplifier gain of 4.7M is needed at this stage. The output is fed to a highpass filter with a cutoff frequency of 0.8Hz whose output is fed to a lowpass

filter aplifier with cuttoff frequency of 4Hz and a gain of 5. This amplified, filtered signal is fed to the microcontroller board. The obtained PPG signal is shown in Figure-8. The photograph of the SpO_2 module setup for testing is shown in Figure-9.

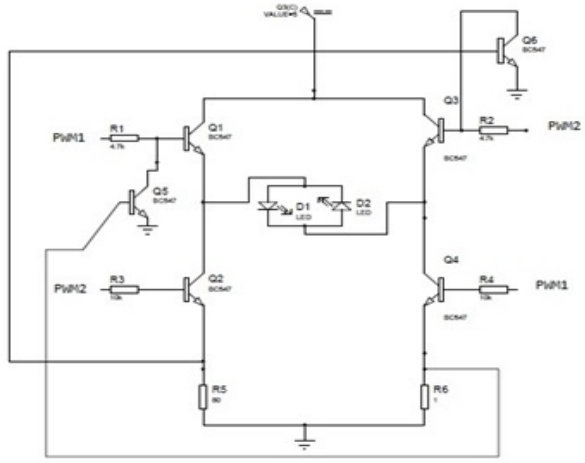

Figure-6. Photo device driving circuit.

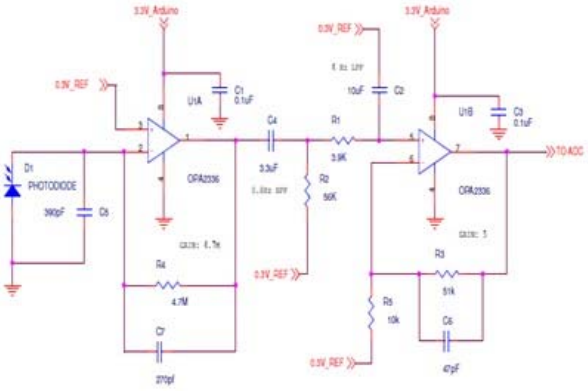

Figure-7. Transimpidance amplifier & filter amplifier.

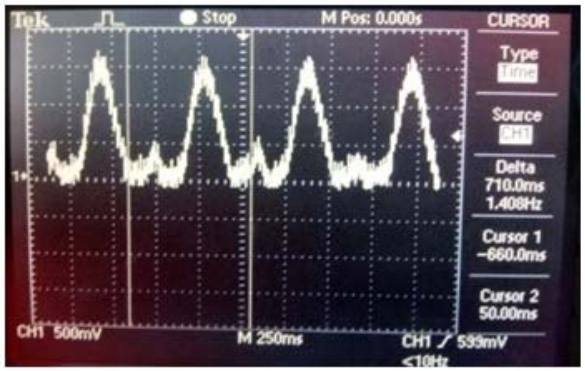

Figure-8. Amplifier output displayed in oscilloscope.

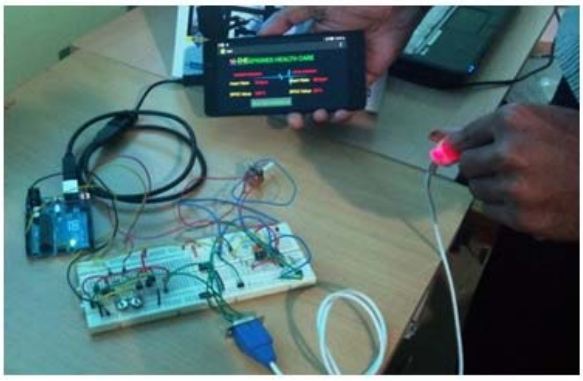

Figure-9. SpO_2 module prototype testing.

DATA ACQUISITION AND CONTROL SOFTWARE

Software part can be divided into two parts. First part is the Microcontroller code and the second part is Android code. The input to the former is the digital signal stream from the inbuilt ADC of the microcontroller. Microcontroller collects the input, continiously stores and create transmission packets and transmit them to the smart phone running Android applications, using an internal protocol that ensures error free transmission. The android applications in the smart phone is divided into background processes for data collection and alert generation and a foreground process for man machine interfaces. Figure-10 shows the background process for data collection. In this process the signal stream sent by microcontroller is collected and stored in the phone's local database after proper time stamping. This is locally processed to compute pulse rate and SpO_2 that are used for further clinical analysis.

The flow diagram for the background process for alert generation and transmission to the cloud server connected to the hospital HIMS system shown in Figure-10. It reads vales of SpO_2 and pulse rate from local database and compares them with the clinician set values. If it falls below the setpoint, it generates an alarm. It also monitores these data streams to find out whether any undesirable changes (COPD or Hypoxemia) is occuring.

DISPLAY SOFTWARE ON SMART PHONE

The user initiated foreground android application displays the instantanious values of Heart Rate and SpO_2 from the local data base. Figure-11 shows main screen of this android application. It has two sections, the current value and the local minimum value. The former gives the instantaneous values of heart rate and SpO_2. The latter displays the minimum value occurred during the current episode.

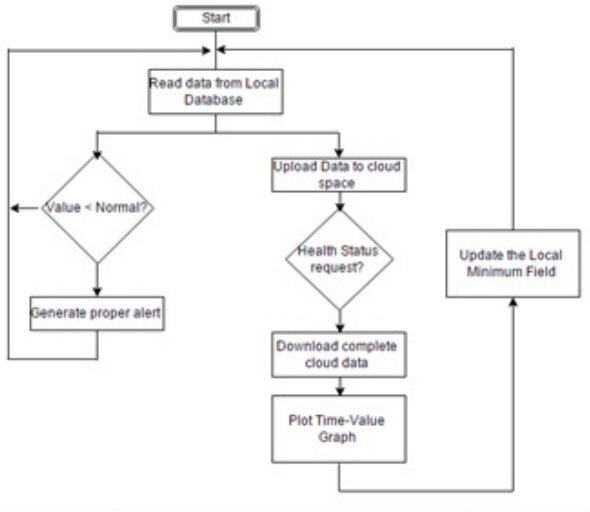

Figure-10. Background process for alarm generation.

Figure-11. Android application main screen.

The button named "Show my Health Status" can be used to switch to the Health status screen as shown in Figure-13. The health status screen can be used to analyze the SpO_2 and pulse rate in a graphical manner. It shows the Time-Value plot of SpO_2 and pulse rate for entire usage time. The zooming options can be used for scaling the time axis.

PROTOTYPE PACKAGING

We have developed an apron based package for the system, the photograph of which is given in Figure-12. The primary aim of this prototype packaging is testing of the concept rather than usage in an actual clinical setting. The main components pertaining to SpO_2 are SpO_2 probe, amplifier and filtering circuit, the Arduino board, the OTG cable and the smart phone used for computational and man machine interface requirements.

Figure-12. Packaging of prototype.

The prototype houses all the subsystems including ECG, Pressure, and Temperature sensors developed for e-PMS. The whole hardware is packaged in a plastic box and an apron with a bag is used to house the system as a wearable monitor.

An android service is set up in the Smart Phone to receive the packets and perform all computations to calculate the pulse rate and SpO_2 values every second as described in section V. The android display applications operate on them to provide human comprehensible display formats similar to the one given in Figure-13.

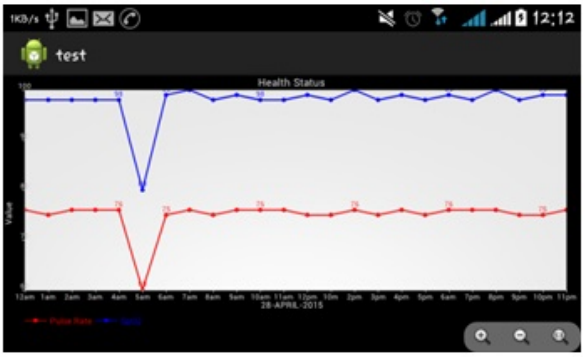

Figure-13. Android application health status screen.

CONCLUSIONS

Technological development in embedded systems is an enabler for smarter ways of monitoring of clinical parameters. The development of a low cost SpO_2 monitoring system that can be easily interfaced to any Android device will be helpful to hasten this process. It will be very effective in providing early detection of hypoxemia and COPD, even in the home healthcare scenario. It may also be used in many applications including training for sportsmen, health condition detection for normal persons, etc.

ACKNOWLEDGEMENTS

This paper is the outcome of a collaborative effort from two institutions -National institute of electronics and information technology (NIELIT) Calicut and Mobilexion Technologies Pvt. Ltd (Mobilexion) Trivandrum. We are deeply indebted to the management of these institutions for having provided a congenial atmosphere for undertaking this work.

REFERENCES

[1]. Vijaylakshmi Kamat. 2002. "Pulse Oximetry". Indian Journal of Anaesthesia.

[2]. William New, Mark Yelderman M.D. "Evalution of Pulse Oxymetry". Anesthesiology V 59, No4, Oct 1983.

[3]. Aravind B, Anil P.C. and Martin K.M. 2014. "Optimal Pre-processing Techniques for a Low-cost Wearable Multi-channel System for Continuous Acquisition of ECG Signals", International Conference on Innovations in Information, Embedded and Communication Systems.

[4]. Mohamed A. Zaltum, M. Shukri Ahmad, Ariffuddin Joret and M. Mahadi Abdul Jamil. 2013. "Design and Development of a portable Pulse Oximetry system". International Journal of Integrated Engineering (IJIE).

[5]. J. Sol`a, S. Castoldi, O. Ch´etelat, M. Correvon, S. Dasen *et al.* 2006. "SpO_2 Sensor Embedded in a Finger Ring: design and implementation". Proceedings of the 28th IEEE EMBS Annual International Conference New York City, USA, Aug 30-Sept 3. ISBN:1-4244-0032-5.

[6]. V. Kimball *et al.* 2004. "Pulse oximeter," United States Patent Office, vol.US006763256B2.

[7]. Zhou Yan and Zhu Jiaxing. 2010. "Design and Implementation of ZigBee Based Wireless Sensor Network for Remote SpO_2 Monitor". Future Computer and Communication (ICFCC), 2nd International Conference. ISBN: 978-1-4244-5821-9.

[8]. Anan Wongjan, Amphawan Julsereewong and Prasit Julsereewong. 2009. "Continuous Measurements of ECG and SpO_2 for Cardiology Information System ".Proceedings of the International Multi Conference of Engineers and Computer Scientists Vol. IIIMECS 2009, March 18 - 20, Hong Kong. ISBN: 978-988-17012-7-5.

[9]. T. Bheema lingaiah, D. Hanumesh Kumar and C. Nagaraja. 2013. "Measurement of Pulse rate and SpO_2 using Pulse Oximeter developed using LabVIEW". IOSR Journal of Electrical and Electronics Engineering (IOSR-JEEE) e-ISSN: 2278-1676,p-ISSN: 2320-3331, Vol. 8, No. 1 (Nov. - Dec.), PP 22-26.

[10]. "SpO Pulse Ox Wrist Oximeter Reference Design". Texas Industries reference note. www.ti.com/lit/ug/tidu124/tidu124.pdf.