

DEVELOPMENT OF WEARABLE PULSE OXIMETRY FOR TELEHEALTH MONITORING SYSTEM

Othman O. Khalifa, T. A. Khan and H. M. Hazzaa

International Islamic University Malaysia
Department of Electrical and Computer Engineering
Jalan Gombak, Malaysia
Khalifa@iium.edu.my

Abstract: In today's healthcare practice there has been increasing interest in wearable/mobile health monitoring devices. These devices allow continuous or intermittent monitoring of physiological signals for the diagnosis as well as treatment of diseases. One of these devices is Pulse oximeter which indirectly measures the oxygen saturation of a patient's blood (as opposed to measuring oxygen saturation directly through a blood sample) and changes in blood volume in the skin, producing a photoplethysmograph (PPG). The main goal of this paper is to demonstrate the overall process involved in the development of a cost-effective user-friendly Pulse Oximetry for Telehealth Monitoring System.

Key words: Pulse Oximetry, Health Monitoring, PPG, light emitting diodes, Photo detectors, oxygen saturation.

1. Introduction

In recent years there has been increasing interest in wearable/mobile health monitoring devices, both in research and industry. These devices are particularly important to the world's increasingly aging population, whose health has to be assessed regularly or monitored continuously. Data derived from sensors can provide input information in models developed to assess and predict the health status of human. Pulse oximeters are commonly employed in nearly all hospital areas where patients are at an increased risk of developing. It is a medical device that indirectly measures the oxygen saturation of a patient's blood (as opposed to measuring oxygen saturation directly through a blood sample) and changes in blood volume in the skin, producing a photoplethysmograph (PPG). It appeared in the early 1980's and gained overnight popularity. It is often attached to a medical monitor so staff can see a patient's oxygenation at all times. It uses optical sensors and light emitting diodes emitting light at different wavelengths through a finger tip where the transmitted light is detected using an optical sensor [1]. Based on the principle of oxygenated haemoglobin having a higher absorption coefficient for infrared light than deoxygenated haemoglobin while deoxygenated haemoglobin absorbs more red lights, by taking the ratio of absorbed red light to infrared light, the oxygen saturation level can be obtained. The waveform has

a complex shape that should be preserved through sensor detection and signal processing.

2. Overview Background

PPG pulse oximetry relies on the fractional change in light absorption due to arterial pulsations. In a typical configuration, light at two different wavelengths illuminating one side of tissue (e.g., a finger) will be detected on the same side (reflectance mode) or the opposing side (transmission mode) after traversing the vascular tissues between the source and the detector [2-11].

Several prominent features can be extracted from the Photoplethysmogram (PPG). Monitoring multiple variables using a single sensor has distinct advantages. The most challenges is to have a very light device that can be operated effectively and efficiently and yet comfortable enough for any user. The possibility of making such a prototype was studied and electronic circuits were developed to take measurements of transmitted light at the two different wavelengths.

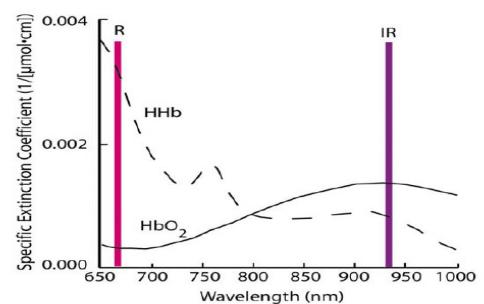


Fig 1. Absorptivity of Hb and HbO₂ in the visible and near-infrared wavelength region [3]

Theoretically, the oximeter has two significant functions that we can use to monitor user health's status. First, the oximeter is used to measure the oxygen saturation (SpO₂) in user arterial and second it measures the user heart rate. The PPG signal is comprised of a non-pulsatile part, referred to as the DC component, and a pulsatile part, the AC component, which are used to calculate an individual's

SpO₂ [2]. The DC component is due to light absorption by skin, tissue, venous blood, and nonpulsatile arterial blood. The AC component is due to light absorption associated with pulsatile arterial blood flow. A normalization technique, where the AC component is divided by the DC component, shown below:

$$\frac{R}{IR} = \frac{AC_r/DC_r}{AC_{ir}/DC_{ir}} \quad (1)$$

The equation eliminates the time invariant absorbance due to venous blood or surrounding tissues. This normalization is carried out for both the red (R) and the infrared (IR) wavelengths. The normalized R/IR results in the so called “ratio of ratios” which pulse oximeter manufactures relate to a set of empirical calibrations specific to their device to obtain SpO₂. There are two distinct methods used to extract HR from the PPG signal [2][3]. The first method identifies each individual heart beat peak or cardiac cycle and determines the peak-to-peak time interval, which corresponds to the time to complete one cardiac cycle. Knowing this interval, HR can be easily calculated as the reciprocal of the time interval. The second method relies on the identification of the frequency of cardiac pulsations. A transform, such as the Fourier transform, is used to describe the energy in the frequency components of the PPG waveform, which results in a power spike at the cardiac frequency.

PPG pulse oximetry relies on the fractional change in light absorption due to arterial pulsations. In a typical configuration, light at two different wavelengths illuminating one side of tissue (e.g., a finger) will be detected on the same side (reflectance mode) or the opposing side (transmission mode) after traversing the vascular tissues between the source and the detector [4]. When a fingertip is simplified as a hemispherical volume that is a homogenous mixture of blood (arterial and venous) and tissue, the detected light intensity is described by the Beer-Lambert law.

3. Related Work

The healthcare has been entering a new stage in the digital age. Previous care delivery models included paid nurse visits, traditional phone-based telehealth applications, and assisted living/nursing home care, each with its own problems. But technological advances are making over the healthcare industry. Digital technology has produced new medical devices such as networked glucose readers, digital thermometers, and stethoscopes, as well as innovative applications such as motion sensors and video-conferencing tools. A pulse oximeter is one of them; it is intended for the non-invasive measurement of arterial blood oxygen saturation and pulse rate [4]. It was initially used for in-hospital use on patients undergoing or recovering from surgery.

During the last few years there has been a significant increase in the number of various pulse oximeters on the market ranging from simple pulse monitors to portable wireless digital pulse oximeters [12][13]. The wireless measurement of oxygen saturation is mostly carried out with finger or ear lobe sensors.

Aoyagi T et al., (1974) invented the fundamental pulse oximetry concepts by describing the use of the arterial pulse for oximetry [13].

There are many research papers since 1974 till now to achieve high efficiency and reliability and reducing cost in pulse oximetry field. In 2009 when we had the first wireless pulse oximetry.

Furthermore, (José M. Castillo et.al) propose prototype to convert a conventional pulse oximeter into a wireless device. And they used zigbee protocol where the data is sent through the wireless network to a central server the system generates alarms to the medical staff when vital signals are critical. And the result of this proposal was accurate and reliable [14]. In contrast a portable real-time wireless health monitoring system have been developed by (N. Watthanawisuth et.al) but the data was transferred within a group of wireless personal area network (WPAN) to database computer server. The system worked successfully and can monitoring the patient health on an area of 10-15 square meters [15].

Webster JG in 2008 have been try to design pulse oximetry which measures the presence of a third substance, carboxyhemoglobin (HbCO), by unfiltered LEDs at 660 nm, 810 nm, and 940 nm. and defines the three absorption equations at three separate wavelengths for nonfocussed LED sources. The result of this design was some erroneous data due to high levels of sulfhemoglobin and methemoglobin and which giving wrong result [16]. Collin schreiner et al. (2010) presents a prototype novel chest based pulse oximetry system and reports on test results from comparative trials with a commercially available finger based pulse oximetry system [17]. Similarly, Adochiei et al. (2011) presented the realization of a wireless low power pulse oximetry telemonitoring system capable to measure and transmit patient’s arterial blood oxygen saturation (SPO₂) level and heart rate (HR) [3].

4. Methodology and Design Process

The block diagram of the methodology of pulse oximeter design is shown in Figure 2. This system consists of two LEDs emitting lights of different wavelengths: infra-red (IR) and red. These LEDs are placed in a box where the finger can be placed. The two LEDs: red LED and infrared LED, and their respective detectors are set in the box. The photo detectors sense transmitted light after being absorbed in the finger. The output of these detectors produces two sets of signals which are then fed into the signal processing block. The signals need to be amplified before it enters a fitting microcontroller which executes further processing in order to calculate ratio of ratios of the signals and compares with threshold value for danger level. This desired parameter is then displayed in a monitor via

wireless implementation.

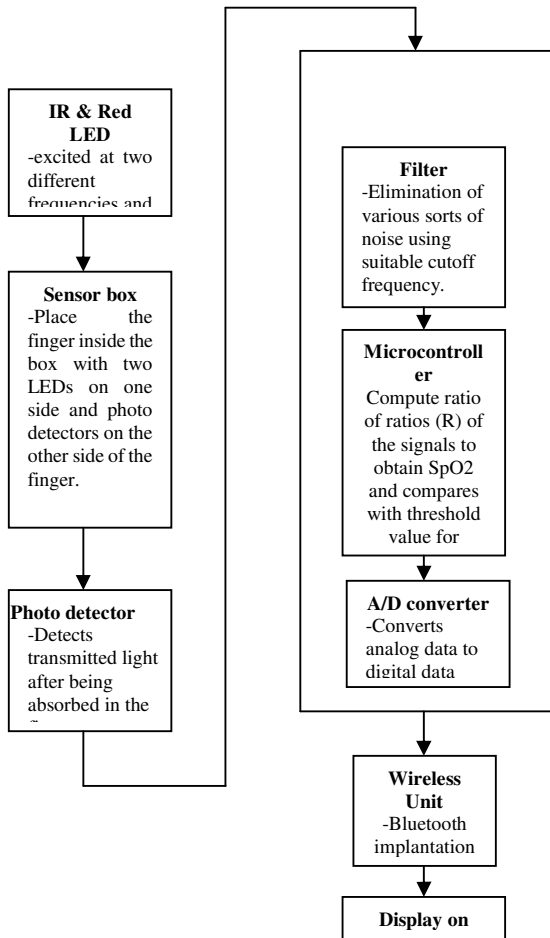


Fig. 2: Scheme for pulse oximeter

5. Hardware Construction and Implementation

The newly designed low-cost Pulse Oximetry hardware structure as shown in Figure 3. It consists of Much of the filtering, amplification, and signal separation. Also, a simple circuit consisting of variable resistors, two LEDs and their respective sensors were constructed. The resistors were included to avoid overloading of the LEDs and the sensors. As mentioned earlier, each LED emits different wavelength: red (660nm) and infra-red (940-950nm). The red light sensor is a light dependent resistor (LDR) whose resistance decreases with increasing incident light intensity and an infrared diode sensor is used as infrared detector. The detected voltage variation is then passed into a low pass filter of 10Hz cutoff frequency.

A filter chip is used to reduce the size of hardware since the aim is to build a portable device. A cutoff frequency of 10Hz was selected as maximum pulse rate of human is 2 Hz. A 10 Hz cutoff frequency is rather high compared to the pulse rate as cutoff frequency of 2 Hz is difficult to achieve. A cutoff frequency of 2 Hz using UAF42Ap would require about 80 MΩ resistor which would act like open circuit thus an impractical choice

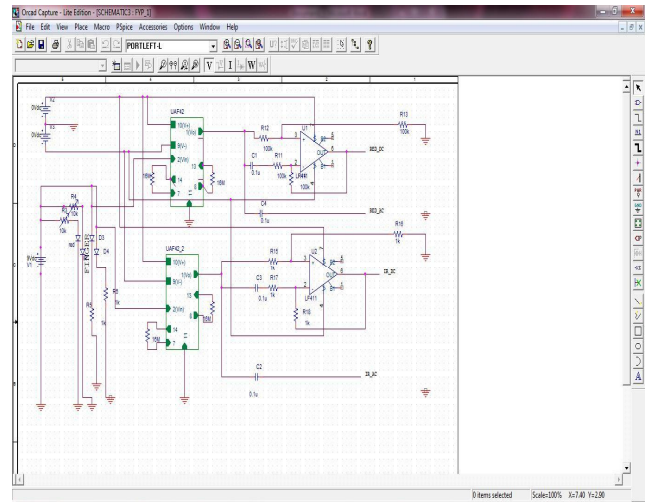


Fig. 3 PSpice schematic of Hardware Construction

The output of low pass filter gives a noise free signal. It is an AC signal with DC offset which is in line with Figure 4. In order to extract the AC component, the signal is passed through a capacitor. A 100nF ceramic capacitor is placed in series with the low pass filter output, which blocks frequency between 0 Hz and 0.5 Hz. The DC component is isolated using a differential amplifier i.e. subtractor. A differential amplifier amplifies the difference between two voltages and rejects the common mode value of the two voltages. The signal with only AC component and signal containing both AC voltage and DC offset is fed into two input terminals of the differential amplifier to obtain the DC component of the signal. Every electrical component was tested individually to verify their respective functionality. The photos of assessment of each component can be seen in Figure 4, 5, 6,.7 and 8.

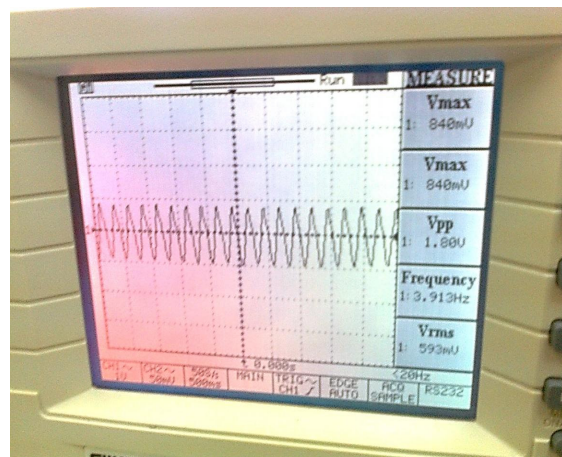


Fig. 4 Functionality testing of low-pass filter prior cutoff frequency

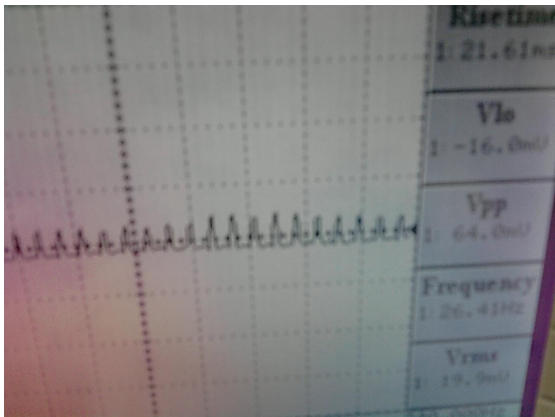


Fig. 5 Functionality testing of low-pass filter after cutoff frequency

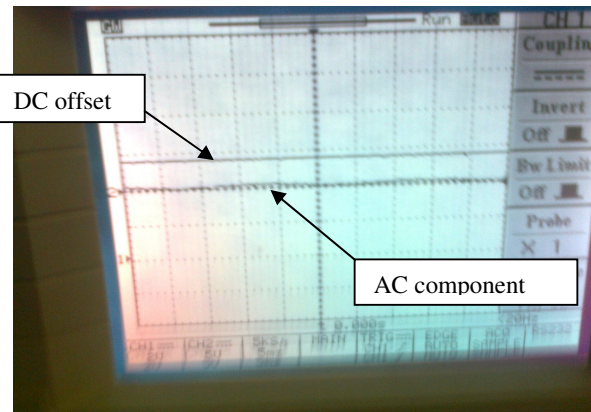


Fig. 8 AC component and DC component for the pulse oximetry signal

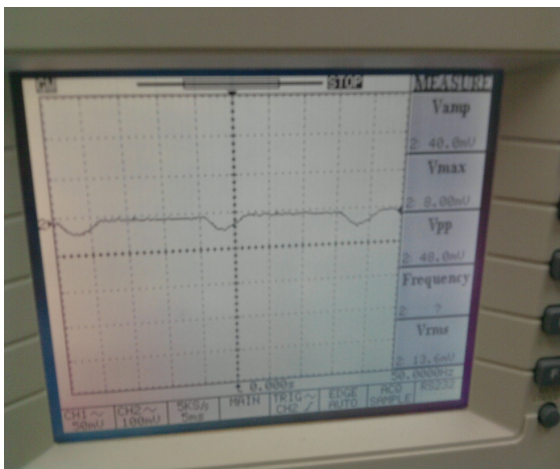


Fig. 6 AC component through capacitor

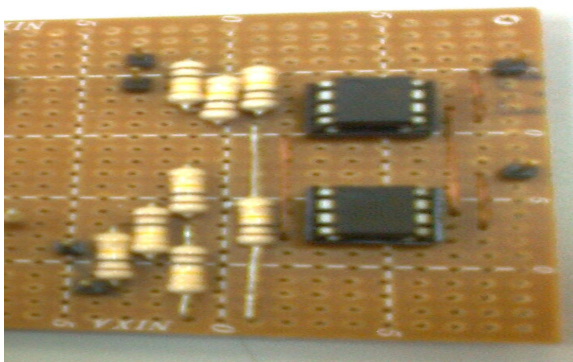


Fig. 7 Implementation of differential amplifier (subtractor)

6. Computing of Pulse Oximetry

The AC component and DC component attained were fed into a microcontroller to calculate the blood oxygen saturation (SpO₂) and pulse rate. An Arduino microcontroller was used to carry out this task. Arduino is an open-source single-board microcontroller, descendant of the open-source wiring platform, designed to make the process of using electronics in multidisciplinary projects more accessible. The hardware consists of a simple open hardware design for the Arduino board with an Atmel AVR processor and on-board input/output support. The software consists of a standard programming language compiler and the boot loader that runs on the board. The microcontroller is programmed using C++ syntax to sense the voltage difference and count occurrence of such voltage variations with time. The code written for Arduino operation calculated blood oxygen saturation and pulse rate simultaneously. For blood oxygen saturation calculation, four signals: AC component from Red LED signal, DC offset obtained from Red LED signal, AC component from IR LED signal and DC offset from IR LED signal, were fed to four selected analogue inputs of the microcontroller. The program began with initiation of variables, assigned pins and pin set up. The microcontroller read signals for 30 seconds in order to get an average value of each signal. The ratio was calculated. The program was also equipped with the SpO₂ calibration graph (Figure 9); this was done by providing coordinates of the SpO₂ calibration graph in a matrix form to be used as look-up table. The calculated 'ratio of ratio' value was then compared with the data set from the look-up table by employing a 'for' loop. Linear interpolation was applied to obtain the corresponding blood oxygen saturation percentage. If the SpO₂ percentage is lower than 90%, a LED would start blinking to alert the patient.

The same program was also used to measure heart rate. The AC and DC components of the red LED sensor were utilized to obtain the pulse rate. One variable is assigned to count the pulse rate while another variable

counts time simultaneously; this is implemented in a loop. This part of the program compared the two signals to count the HIGH of the pulsatile signal. The summation of the number of 'HIGH' over one minute is the pulse rate. To accommodate both SpO2 percentage and pulse rate, the signals were read for 30 seconds, therefore, the final answer of the pulse rate is twice the value obtained in 30 seconds.

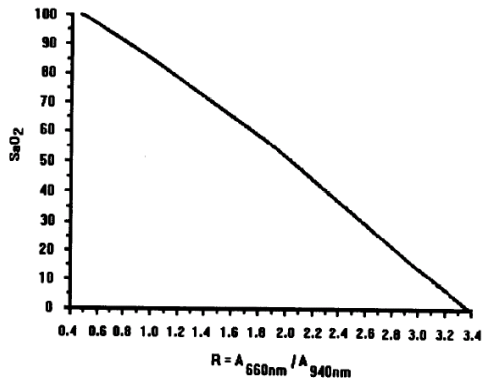


Fig. 9. Empirical Oxygen Saturation
(Curve Tremper, Boston, Little & Company, 1987)

7. Results and Discussion

The designed pulse oximeter was tested on several people of different age, gender and ethnicity. The results are tabulated in Table 1 and Table.2. It observed that system was able to produce highly reliable results for both SpO2 and Heart Rate.

Sub ject	Age	Gender	SpO2(%)	Heart Rate (beats/min)
1.	22	F	95.97	67
2.	22	F	96.07	71
3.	23	F	96.99	74
4.	30	M	97.13	75
5.	30	M	95.87	80
6..	33	M	96.32	77

Table 1 Pulse oximeter testing in normal conditions

The readings shown in Table 2 were taken in normal conditons i.e. the subjects were at ease and breathing normally. The SpO2 percentatge and heart rate readings are

inline with the theory. The blood oxygen saturation of all subjects are above 95% as it is suppose to be for an healthy person. In order to test the funtionalty of the device, another set of readings was carried out wherby the subjects were asked to hold their breath for 30 seconds. The obtained result are shown in Table 3.

Subject	Age	Gender	SpO2(%)	Heart Rate (beats/min)
1.	22	F	93.13	82
2.	30	M	90.24	77
3.	22	F	89.77	80

Table 2: Pulse oximeter testing with low oxygen supply

This task proved to be rather challenging as no one was able to hold their breath for 30 seconds continuously. It became more difficult to measure one person repeatedly as he/she would suffer from breathlessness. However, the readings do indicate low blood oxygen satuation level as they are lower than 95%, eventhough only by small amount. Blood oxygen saturation percentage ranging from 90-94% indicate very mild hypoxia which is not hazardous if not recurring. This experiment proved that the pulse oximeter is indeed functioning properly. In order to verify the accuracy of the device, the heart rate measured by the device was compared to heart rate counted using stethoscope. The authenticity of the device was tested utilizing pulse rather than SpO2 level due to unavailabilty of pulse oximeters in the market compared to stethoscope. The comparison is tabulated in Table 3.

Subject	Wireless Pulse Oximeter	Stethoscope
1	66	66
2	78	80
3	69	72
4	83	83
5	69	70

Table 3 Performance testing of Pulse oximeter

Table 3 shows that the stethoscope result is always higher than that of the heart rate monitoring system; this could be due to shortcomings of the system which may have missed few pulses. Pulses can be missed due to its high speed or insufficient amplification which wasn't able to meet the threshold. Inaccuracy could also rise from human error whereby mistake in counting of pulse might have occurred. However, the error percentage is not very large; it

is lower than 5%. This comparison confirmed the performance of the device in whole, as blood oxygen saturation measurement and pulse rate use same infrared signal.

8. Conclusion

There is no doubt that in the context of wearable medical devices, the pulse oximetry represents the greatest advance in wearable patient monitoring in many years. It is a medical device that monitors the level of oxygen in a patient's blood and alert if oxygen levels drop below safe levels allowing rapid intervention. This work built a Bluetooth based pulse oximetry monitoring system. A simulation was performed. The results were obtained through simulation forecasted the likely difficulties that may take place during the hardware implementation. Hardware was designed to compute blood oxygen saturation as well as pulse rate to determine state of a patient. The signals produced by the two different wavelength absorption had noise added to them. Active filters were implemented to process the signals and eliminate noise. AC component and DC component of each signal were extracted by employing capacitor and differential amplifier. Arduino microcontroller was used to calibrate and calculate the SpO₂ percentage and pulse rate simultaneously. The computed result was displayed on a laptop via Bluetooth implementation. A condition was executed in the code to trigger alarm in the form of blinking LED if SpO₂ percentage was lower than safe level of 90%.

References

1. Valdez-Lowe, C.; Ghareeb, S.A., Artinian, N.T. "Pulse oximetry in adults". *Amer J Nrs* 2009; 109(6), 52-59.
2. Ferrera, F., Ferrero, F., Blanco, C., Viera, J., Vega, M., Blanco, J. (2006). Design of a low-cost instrument for pulse oximetry. *Proceedings of the IEEE Instrumentation and Measurement Technology Conference*, pp. 573-577.
3. H.costin,C. rotariu ct al., (2009), complex system for real time medical telemomitoring of vital signs ,ISSN:2066-7590,PP.17-23..
4. John W. Severinghaus, MD, "Takuo Aoyagi: Discovery of Pulse Oximetry", *Anesthesia & Analgesia*, vol. 105 no. 6S Suppl S1-S4 2007.
5. Kamat V., "Pulse Oximetry", *Indian Journal of Anaesthesia*, 2002, 46 (6), 261-268.
6. MacDonald, M., (2009). "Your Body: The Missing Manual. Sebastopol", CA: Pogue Press. ISBN 0-596-80174-2.
7. Mardirossian G. and Schneider, R.E., "Limitation of Pulse Oximetry", *Anesth Prog*. 1992; 39(6): 194-196.
8. Marinos, D., Vlisidis, N., Giovanis, C., Leonidas, F., Vassilopoulos, C., Aidinis, C., Pagiatakis, G., Schmitt, N., Klaue, J., Pistner, T., "Passenger health monitoring applications for aircraft cabin environment". *International Conference for scientific computing to computational engineering, IC-SCCE Athens* june 2008.
9. Parlato, M.B., Meyer, J., Dzyubak, B., Helfenberger, J., "Low Cost Pulse Oximeter Probe", In *Conjunction with Engineering World Health and the MedeCal Project*, 2009.
10. Adochiei, F.; Rotariu, C.; Ciobotariu, R.; Costin, H.; , "A wireless low-power pulse oximetry system for patient telemonitoring," *Advanced Topics in Electrical Engineering (ATEE)*, 2011 7th International Symposium on , vol., no., pp.1-4, 12-14 May 2011.
11. Anastasopoulos, K.; Kouros, P.; Marinos, D.; Aidinis, K.; Schmitt, N.; Klaue, J.; Pistner, T.; , "Portable embedded medical box implementation for health monitoring applications in avionic environments," *Medical Measurements and Applications*, 2009. MeMeA 2009. IEEE International Workshop on , vol., no., pp.17-21, 29-30 May 2009 doi: 10.1109/MEMEA.2009.5167946.
12. Aoyagi T, Kishi M, Yamaguchi K, Watanabe S. (1974), "Improvement of the earpiece oximeter ,In: Abstract of the 13th annual meeting of the Japanese society for the medical electronics and biological engineering .p.99-91.
13. Castillo, J.M., Olivares, J., Palomares, J.M." Design of a Wireless Pulse Oximeter using a Mesh ZigBee Sensor Network" , *Department of Computer Architecture, University of Córdoba, Spain, Bio Devices*, 2011, pp 410-404.
14. Watthanawisuth, N.; Lomas, T.; Wisitsoraat, A.; Tuantranont, A.; , "Wireless wearable pulse oximeter for health monitoring using ZigBee wireless sensor network," *Electrical Engineering/Electronics Computer Telecommunications and Information Technology (ECTI-CON)*, 2010 International Conference on , vol., no., pp.575-579, 19-21 May 2010.
15. Pulse Oximetry in the Patient's Hands, <http://www.rtmagazine.com/2014/02/pulse-oximetry-patients-hands/>. Retrieved October 2014.
16. A.Keerthika,R.Ganesan, Pervasive Health Care System for Monitoring Oxygen Saturation using Pulse Oximeter Sensor, *Proceedings of 2013 IEEE Conference on Information and Communication Technologies (ICT 2013)*, pp. 819-823.