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Development of a Stand-Alone Pulse Oximeter

Master Thesis

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Electronics and Instrumentation Group
Instrumentation Center
Physics Department - FCTUC

Intelligent Sensing Anywhere

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Student Number: 2004107061

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ABSTRACT

Nowadays, the biomedical instrumentation holds a prominent position within medicine. Following this trend, the pulse oximeter has become an important tool to elucidate about the functioning of the organism and wakeup for anomalies by monitoring the heartbeat and the level of oxygen saturation in the blood that flows in the human body. These devices are mostly used in hospitals and clinics but are gradually finding their way into domestic use.

The goal of this thesis is to develop and test technology solutions to implement a cheap, accurate, reliable and easy to use finger transmission pulse oximeter probe able to interface an acquisition module and then data is sent to a processing module that will process the signals.

Electronic circuits were developed to take measurements of light transmitted through the finger at two different wavelengths. These circuits include three functional modules of the oximeter probe which allow the signal acquisition: a LED driver module which controls the amount of drive current; a photodetection module which detects light that is transmitted through the finger and converts the electrical signal into voltage; and a timing module which allows the LEDs to switch in order to light up alternately.

The obtained results show that the oximeter probe developed has a good performance and is able to detect transmitted light through a human finger with variations in the amplitude of voltage at two wavelengths, making possible to calculate the percentage of oxygen saturation in the blood and simultaneously the heart rate.

RESUMO

Hoje em dia, a instrumentação biomédica ocupa posição de destaque dentro da medicina. Seguindo essa tendência, o oxímetro de pulso tornou-se num importante instrumento para elucidar acerca do funcionamento do organismo e alertar para eventuais anomalias através da monitorização do batimento cardíaco e do nível de saturação de oxigénio do sangue que flui no corpo humano. Estes equipamentos são maioritariamente encontrados em hospitais e clínicas mas gradualmente estão a ser cada vez mais utilizados a partir de casa.

O objectivo desta tese é desenvolver e testar soluções tecnológicas para implementar uma ponta oximétrica de transmissão para o dedo barata, exacta, durável e fácil de usar que possa ser capaz de interagir com um módulo de aquisição e posteriormente os dados são enviados para um módulo de processamento que efectuará o processamento dos sinais.

Foram desenvolvidos circuitos electrónicos para efectuar medidas da luz transmitida através do dedo a dois comprimentos de onda diferentes. Esses circuitos incluem três módulos funcionais da ponta oximétrica que permitem a aquisição do sinal: um módulo de driver dos LEDs que controla a quantidade de corrente no circuito; um módulo de fotodetecção que detecta a luz que é transmitida através do dedo e converte o sinal eléctrico em tensão; e um módulo de temporização dos LEDs que possibilita a comutação destes para que acendam alternadamente.

Os resultados obtidos mostram que a ponta oximétrica desenvolvida tem um bom desempenho e consegue detectar luz através do dedo humano com variações na amplitude de tensão para os dois comprimentos de onda, tornando assim possível o cálculo da percentagem de saturação do oxigénio no sangue e simultaneamente o batimento cardíaco.

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To my family and friends

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

ACRONYMS AND DEFINITIONS

2,3-DPG	2,3-diphosphoglyceric acid
A	Ampere (electric current unit)
A/W	Ampere by Watt
AC	Alternate Current
Bit	Binary Digit
BPM	Beats Per Minute
cm	Centimetre (unit of length)
CPU	Central Processing Unit
DAQ	Data Acquisition
DC	Direct Current
EC	European Community
F	Farad (unit of capacitance)
FDA	Food and Drug Administration
FET	Field Effect Transistor
GEI	Electronics and Instrumentation Group
Hb	Deoxygenated Hemoglobin
HbO_2	Oxygenated Hemoglobin
Hz	Symbol of Hertz (frequency unit)
I/O	Input/Output
I^2C	Inter Integrated Circuit
IC	Integrated Circuit
INFARMED	National Authority of Medicines and Health Products (Portugal)
IR	Infrared
ISA	Intelligence Sensing Anywhere
ISO	International Organization of Standardization
kΩ	Kilohm (electrical resistance unit)
LabView	Laboratory Virtual Instrument Engineering Workbench
LCD	Liquid Crystal Display
LED	Light Emitting Diodes

ln	Logarithm to the base *e*
mA	miliampere (electric current unit)
Matlab	Matrix Laboratory (Software Application Analysis)
MHz	Megahertz (frequency unit)
mm	Milimetre (unit of length)
mm^2	Squared Milimetre (unit of area)
mm^3	Cubic Millimetre (unit of volume)
mV	Milivolt (unit of electromotive force)
NAND	Not AND
nF	Nanofarad (unit of capacitance)
NI	National Instruments
nm	Nanometre (unit of length)
O_2	Molecule of Oxygen
ºC	Degree Celsius (unit of temperature)
OP-AMP	Operational Amplifier
PC	Personal Computer
PCB	Printed Circuit Board
pF	Picofarad (unit of capacitance)
pH	Potential of Hydrogen
PNP	Positive-Negative-Positive
QSR	Quality System Regulation
R	Red
R&D	Research & Development
R_{OS}	Ratio of Ratios
S_aO_2	Oxygen Saturation on Arterial Blood
SMD	Surface-Mount Technology
SPI	Synchronous Serial Interface
S_PO_2	Saturation of Peripheral Blood Oxygen
TTL	Transistor-Transistor Logic
UART	Universal Asynchronous Receiver and Transmitters
US	United States (of America)

USB	Universal Serial Bus
V	Volt (unit of electrostatic potential)
V+	op-amp positive input voltage
µA	Microampere (electric current unit)
Ω	Symbol of Ohm (electrical resistance unit)

1. INTRODUCTION

1.1 Motivations

The measure of the oxygen saturation of a patient's hemoglobin (Hb) in some parts of the circulatory system can give important information about the state of vital organs as heart and lungs and the perfusion in other ones. So, the technique of oximetry can elucidate about the functioning of the organism and wakeup to possible anomalies.

The regular values of the blood oxygen saturation are around 97% [2] and significant changes on those values can be associated to alarming situations. Therefore the technique of monitoring of the oxygen saturation on blood has a wide range of medical applications; particularly, in patients at risk of respiratory failure, it is important to have a measure of the efficiency of the work performed by the lungs and it can be done through the monitoring of how well the arterial blood is oxygenated. The techniques of oximetry play also an important role in the investigation of sleep disorders.

The development of devices based on non-invasive techniques becomes quite important due to some limitations associated of the measures on arteries of the oxygen saturation such as the impossibility of a continuous monitoring and the loss of blood. So this work will focus the technique of pulse oximetry based on the transmission of light in tissues as a non-invasive optical way of monitoring the oxygen blood saturation.

1.2 Objectives

The main propose of this project is to develop a stand-alone oximeter which allows the monitoring of oxygen saturation in a non-invasive way. To that, it is intended to develop an oximeter probe prototype to acquire the biological signals and then a acquisition and processing module integrates the data (through the implementation of algorithms) in order to estimate the oxygen saturation.

The oximetry principles are well studied being described on the literature and are already truly rooted in modern healthcare with a remarkable credibility. So the oximetry solution of this project will not for sure revolutionize the world: what's at stake is not to create an entire new device but "just" optimize the actual knowledge using off-the-shelf

data analyze and hardware components. The purpose is to develop a reliable, low cost and portable oximetry device ready for clinical and domestic use.

It was also proposed interconnect the pulse oximeter with ISA projects, where the device would be used to help in the monitoring of the vital signals. ISA (Intelligent Sensing Anywhere) is a spin-off company of the University of Coimbra that was founded in 1990. The technologic based company integrates a R&D unit that works on the development of complete solutions, which include hardware, firmware and software, for a wide range of application areas, including healthcare.

Particularly, this work will focus on the development of an oximeter probe which will be responsible by the acquisition of the signal.

1.3 Developed Works

The present project is part of a group of works of the Electronics and Instrumentation Group (GEI) of the Instrumentation Center, in the development of instrumentation for the vital signals monitoring.

On the project of "Development of a Stand-Alone Pulse Oximeter", it was very important a previous work developed on GEI, in 1995 by Eng. Rita Jorge de Sousa Costa Pereira, called "Projecto de um Sistema Digital de Medida para Aplicações Biomédicas" [10]. This work provided some background knowledge about the operating of pulse oximeters and helped the students to make some options adequate to the desired work based on the study that had been already done.

1.4 Document Structure

The present thesis has been prepared in six chapters:

- In Chapter 1, the document is contextualized and objectives and motivations are focused;
- In Chapter 2 it will be exposed the theoretical and technical principles of the pulse oximetry required to the beginning of the project development and the state of the art of oximetry, pulse oximeters and oximeter probes. On this

chapter, it is also possible to find information about the legal procedure to commercialization a medical device such as a pulse oximeter in Europe and US;

- Chapter 3 describes the overall system architecture and particularly focused the three different modules of the oximeter probe and the hardware design;
- Chapter 4 shows the obtained results and the contextualized discussion of them;
- Chapter 5 ends the project report with an analysis of actual status of the project, the suggestions for a future work and the student's final appreciation.

2. THEORETICAL AND TECHNICAL BACKGROUND

It was really important to the project the acquisition of theoretical and technical background in order to be possible to accomplish it. So the first step of the project was the acquisition of new concepts and review of other ones.

The following chapter present a general description of the research work carried out during the development of the project for a better technical understanding of the issues of other chapters.

2.1 Study of Hemoglobin

Pulse oximetry is a non-invasive method that allows the monitoring of oxygen saturation of a patient's hemoglobin (Hb). The hemoglobin (Figure 1) is a protein which is present in the red cells of the blood and it is responsible for the oxygen (O_2) transport throughout the body.

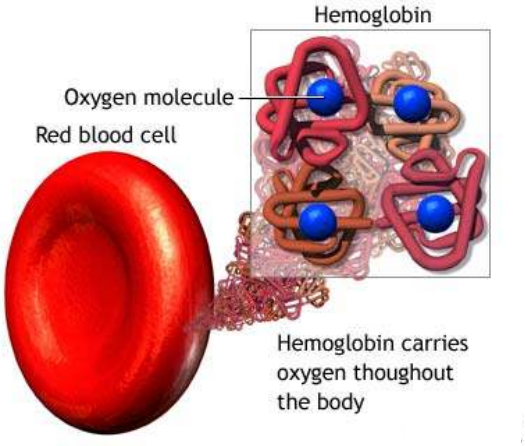


Figure 1. *Structure of hemoglobin.* From [5]

Each hemoglobin molecule is constituted by four polypeptide chains called globins and four disc-shaped organic pigment molecules called hemes. In the center of each heme group, there is one atom of iron which can combine with one molecule of oxygen. So, one hemoglobin can carry four molecules of oxygen [2].

There are two forms of hemoglobin in blood: oxygenated hemoglobin (HbO_2) and deoxygenated hemoglobin (Hb). Hb combines with the oxygen to form HbO_2 in the lungs through a loading reaction, as it is possible to see in Figure 2.

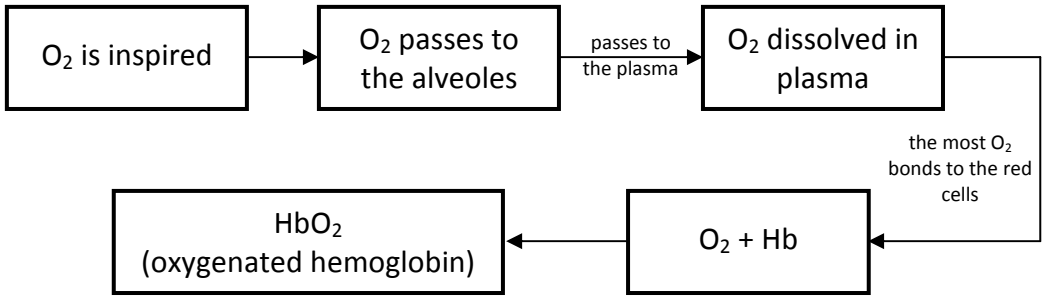


Figure 2. *Schematic of gas exchange in pulmonary capillaries.* Based on [2]

The HbO_2 is transported to the tissues capillaries where it dissociates to yield Hb and free O_2 molecules through an unloading reaction, and oxygen is used in mitochondria (Figure 3).

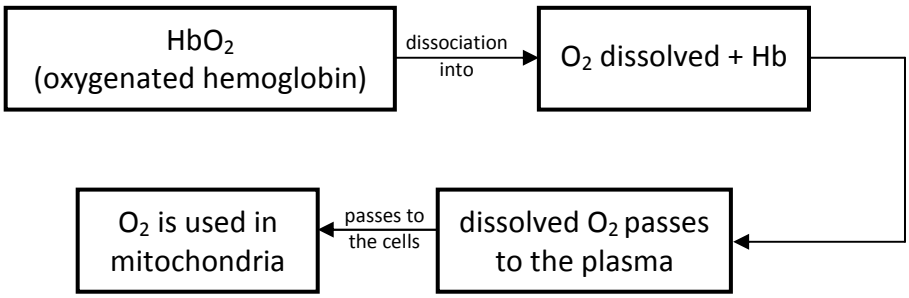


Figure 3. *Schematic of gas exchange in tissues capillaries.* Based on [2]

Loading and unloading reactions of hemoglobin depend on two factors [2]:

- the partial pressure of oxygen (P_{O_2}): in the pulmonary capillaries, there is a high value of P_{O_2} and almost all Hb molecules combine with O_2; in the tissues capillaries a low value of P_{O_2} promote the dissociation of HbO_2;
- the affinity between hemoglobin and oxygen: a very strong bond would favor the loading reaction while a weak bond would hinder the dissociation; the affinity depends on several factors such as temperature, pH and 2,3-diphosphoglyceric acid (2,3-DPG).

Shortly after the discovery of this protein (in 1860 [6]), it was concluded that the absorption of light by hemoglobin varies according the saturation in oxygen. When the hemoglobin without oxygen bond with the oxygen to form oxygenated hemoglobin, it becomes red; in the dissociation of oxygen hemoglobin gets darker. This difference of

colours is because Hb and HbO_2 have also a difference in the optical spectral in the range of wavelengths between 600nm (close to red) and 1000nm (near infrared) [1], [2].

So pulse oximetry is mainly based on the characteristics of absorption of red and infrared light by the oxygenated and deoxygenated hemoglobin: the oxygenated hemoglobin absorbs more infrared light and transmits more red light, while the deoxygenated one absorbs more red light and transmit more infrared light, which can be seen in Figure 4:

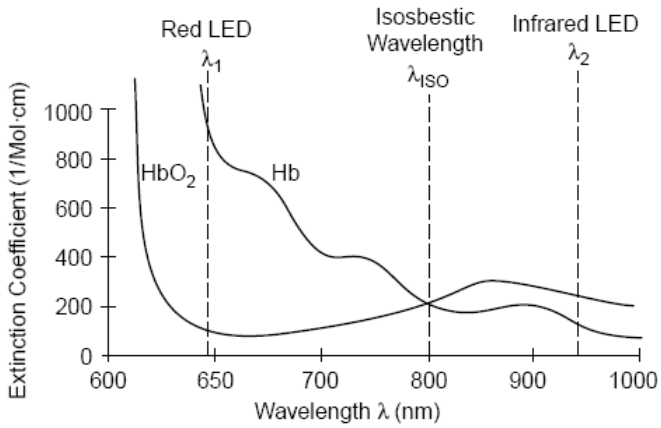


Figure 4. *Graph of the absorption curves for both types of hemoglobin (oxygenated and deoxygenated) as a function of wavelength.* From [6]

As it is possible to see in Figure 4, oxygenated and deoxygenated hemoglobin have a significantly different optical spectra in the wavelength range from 600nm to 1000nm. The difference is big in some wavelengths (around 660nm in the red region and around 910nm in the infrared region) and small or not existing in other ones (isobestic wavelength). The difference in these two wavelengths can be used to calculate the oxygen saturation in blood.

However, these two forms of hemoglobin are not the only one that exists in the patient's blood. There are other abnormal hemoglobins such as carboxyhemoglobin[1] and methemoglobin[2] and each one of them also absorbs light and has its own extinction coefficient curve as it is possible to see in Figure 5 [6].

[1] hemoglobin that has carbon monoxide instead of the oxygen bound to it
[2] hemoglobin whose the iron is in the Fe^{3+} state, not the Fe^{2+} of normal hemoglobin

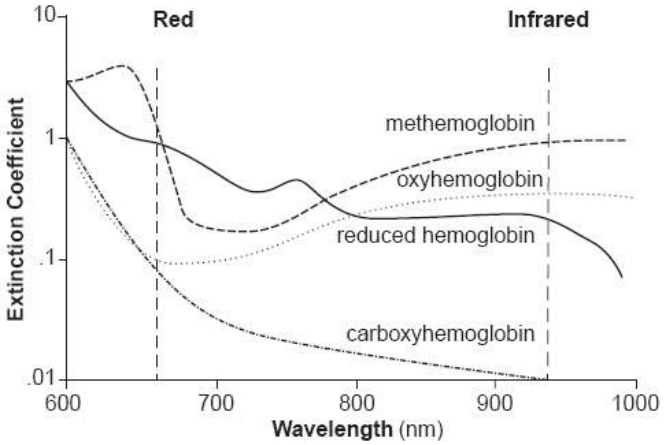


Figure 5. *Absorption spectra of common forms of haemoglobin (oxyhemoglobin, deoxyhemoglobin, methemoglobin, carboxygenoglobin).* From [6]

In Figure 5, it is clear that some of these spectra are very close to the oxygenated hemoglobin spectra at the routinely used two wavelengths, which may lead to erroneous and false readings. So, when a patient has a suspicion of high level of carboxyhemoglobin and methemoglobin, it is recommended to use a special oximeter called CO-oximeter [6]. CO-oximetry is an in vitro method of oximetry that uses four or more light wavelengths so it is able to measure the abnormal hemoglobins as well as normal hemoglobins. The oxygen saturation is calculated through samples of blood in a cuvette and it is obtained a direct measurement of the oxygen content of the arterial blood (S_aO_2) [42]. Despite of the benefit of a more accurate result, this method have the inconvenient of being invasive and only give the information for the moment the sample was taken not suitable for continuous monitoring.

The pulse oximetry is also based in another physical principle: plethysmography principle where the absorbance of both wavelengths has a pulsatile component, which is due to the fluctuations in the volume of arterial blood at the sensor site which will trigger to changes in the light transmitted through the tissues [8].

Thus, combining two technologies of spectrophotometry and optical plethysmography mentioned above, a pulse oximeter can provide important information such as the heart rate and the oxygen saturation on the blood of peripheral capillary (S_pO_2), which is an indirect measurement of the oxygen content of blood and represents an estimative of S_aO_2.

2.2. Oxygen Saturation and the Absorption of Light in Tissues

In pulse oximetry, the oxygen saturation in blood (S_PO_2) is the ratio between the concentration of oxygenated hemoglobin and all the hemoglobin present in blood which can be defined by the following equation [1]:

$$S_PO_2 = \frac{[HbO_2]}{[HbO_2]+[Hb]} \text{ x } 100\ (\%) \qquad \textbf{(Equation 1)}$$

where $[HbO_2]$ is the concentration of oxygenated hemoglobin and [Hb] is the concentration of the deoxygenated form.

Beer-Lambert's Law and the Ratio of Ratios (R_{OS}) [1], [4]

The detection of oxygen saturation of hemoglobin is done by spectrophotometry and it is based on Beer-Lambert law which relates the concentration of a solute to the intensity of a monochromatic light transmitted through a homogeneous solution not disperser [1]:

$$I_{trans} = I_0.e^{-\varepsilon(\lambda)CD} \qquad \textbf{(Equation 2)}$$

where:

- I_{trans} is the intensity of transmitted light
- I_0 is the intensity of incident light
- $\varepsilon(\lambda)$ is the extinction coefficient of solute (which depends of the solute and the wavelength used)
- C is the concentration of solute
- D is the optical path distance

Beer-Lambert's law describes the attenuation of light which passes through a medium containing an absorbing solute: once the intensity I_0 focused the medium, part of the light is absorbed and the other transmitted, so intensity I_{trans} is transmitted and it decreases exponentially with the distance traveled by light through the middle [3].

Using the law of Beer-Lambert to measure oxygen saturation in blood, it is necessary to take into account two important factors:

- due to reflection and scattering of light, it is not easy to determine the precise intensity of the incident light applied;
- as the volume of blood at the sensor site varies with the arterial pulse (due to systole[1] and diastole[2]), the thickness of that place also varies slightly with each pulse, because the physical diameters of the arteries increase and decrease periodically due to pressure; therefore, there will be fluctuations in the distance traveled by light that is transmitted.

So the Beer-Lambert's law needs to be modified to eliminate the factors mentioned above and become possible the estimation of oxygen saturation.

Thickness fluctuations caused by arterial pulse can be seen as a change of distance D of the Beer-Lambert's equation (Equation 2).

The human body is not composed by just one component with a concentration C at one absorptivity ε and the intensity of the light transmitted is a function of the absorbance coefficient of both fixed elements (bone, tissue, skin and hair) as well as variable ones (volume of blood). So the $\varepsilon(\lambda)$ term and the *C* term can be lumped together in one term $\alpha(\lambda)$ as a function of wavelength:

$$\alpha(\lambda) = \varepsilon(\lambda).C \qquad \textbf{(Equation 3)}$$

Assuming that a pulsation's minimum provides a baseline intensity component I_1, Beer-Lambert's law can be written as follows:

$$I_1 = I_0.e^{-\alpha_1(\lambda).D} \qquad \textbf{(Equation 4)}$$

Likewise, a pulsation's maximum provides a intensity of light I_2 emerging from the pulsatile component and it is a function of its light intensity I_1 so Equation 1 can be written as a variation of the baseline component set in Equation 4:

$$I_2 = I_1.e^{-\alpha_2(\lambda).\Delta D} = I_0.e^{-(\alpha_1(\lambda).D+\alpha_2(\lambda).\Delta D)} \qquad \textbf{(Equation 5)}$$

where ΔD is the changing thickness of the place of measurement.

[1] where the arterial blood volume is greatest
[2] where the arterial blood volume is lowest

A change in transmission (ΔT) can be defined by taking the relationship between I_1 and I_2 as follows:

$$\Delta T = \frac{I_2}{I_1} = \frac{I_0.e^{-(\alpha_1(\lambda).D+\alpha_2(\lambda).\Delta D)}}{I_0.e^{-\alpha_1(\lambda).D}} = e^{-\alpha_2(\lambda).\Delta D} \qquad \textbf{(Equation 6)}$$

With Equation 6 it was possible to eliminate the input light intensity as a variable. However, the equation is still a function of ΔD, which is impossible to measure. To simplify the equation, the natural logarithmic is taken for both sides of Equation 6 yielding the following:

$$- \ln(\Delta T) = - \ln e^{-\alpha_2.\Delta D} = \alpha_2(\lambda).\Delta D \qquad \textbf{(Equation 7)}$$

The term ΔD may be dropped by measuring the arterial transmission at two different wavelengths. In a pulse oximeter, it is selected one red (*R*) and one infrared (*IR*) wavelengths (λ_R ,λ_{IR}), which are in a range away from the approximate isobestic wavelength that is sufficient to allow the two signals to be easily distinguish:

$$\ln(\Delta T_R) = - \alpha_2(\lambda_R).\Delta D \qquad \textbf{(Equation 8)}$$

$$\ln(\Delta T_{IR}) = - \alpha_2(\lambda_{IR}).\Delta D \qquad \textbf{(Equation 9)}$$

Assuming that the two sources are positioned at approximately the same distance from the photodetector, the term ΔD are the same in Equations 8 and 9. For this reason, ΔD may be eliminated through the following quotient:

$$\frac{\ln(\Delta T_R)}{\ln(\Delta T_{IR})} = \frac{-\alpha_2(\lambda_R).\Delta D}{-\alpha_2(\lambda_{IR}).\Delta D} = \frac{\alpha_2(\lambda_R)}{\alpha_2(\lambda_{IR})} \qquad \textbf{(Equation 10)}$$

Equation 10 is independent of term ΔD but does not give an accurate measurement of oxygen saturation in blood, so it is relied to produce a variable to calculate oxygen saturation. If the ratio of arterial absorbance at the red and infrared wavelengths can be determined, the oxygen saturation of the blood can be calculated using empirical derived calibration curves, independently of I_0 and ΔD.

So, from Equation 10 is defined as the Ratio of Ratios (R_{os}):

$$R_{OS} = \frac{\ln(\Delta T_R)}{\ln(\Delta T_{IR})} = \frac{\alpha_2(\lambda_R)}{\alpha_2(\lambda_{IR})} \qquad \textbf{(Equation 11)}$$

The ratio of Equation 11 is used to calculate the oxygen saturation of the patient's blood.

In the opposite side of both LEDs there is a photodetector that receives the light which is transmitted through the place of measurement. This photodetector picks up two signals, one for the R and the other for the IR, as it is schematically represented in the following graphs in Figure 6:

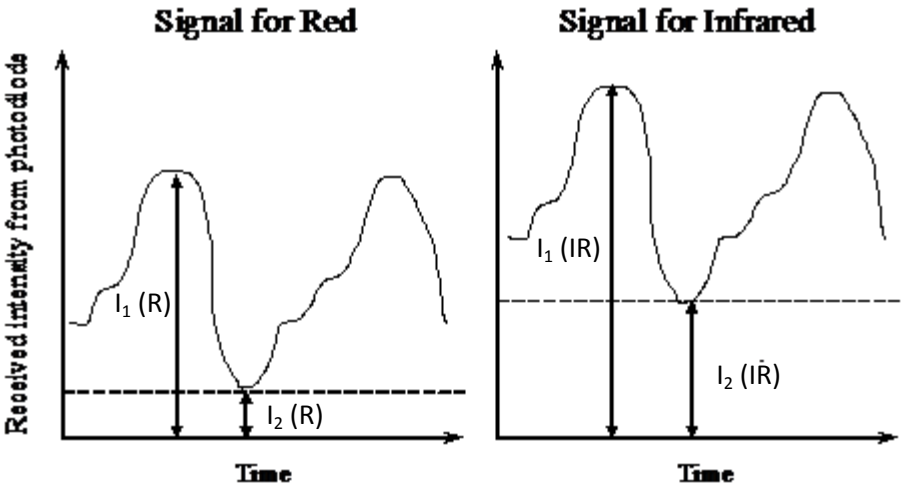


Figure 6. *Graph of transmitted light intensity through the finger: high (H) and low (L) signals as a function of time of the transmission of red (R) and infrared (IR) light. Adapted from* [1]

The received red wavelength varies with each pulse and has high and low values I_1(R) and I_2(R), respectively. The same occurs with infrared light to I_1(IR) and I_2(IR), respectively.

So using Equation 6, a change in transmission can be calculated at each of the two wavelengths:

$$\Delta T_R = \frac{I_2(\lambda_R)}{I_1(\lambda_R)} \qquad \textbf{(Equation 12)}$$

$$\Delta T_{IR} = \frac{I_2(\lambda_{IR})}{I_1(\lambda_{IR})} \qquad \textbf{(Equation 13)}$$

The logarithmic is taken for both sides of Equations 12 and 13 yielding the following:

$$\ln(\Delta T_R) = \ln(\frac{I_2(\lambda_R)}{I_1(\lambda_R)}) \quad \textbf{(Equation 14)}$$

$$\ln(\Delta T_{IR}) = \ln(\frac{I_2(\lambda_{IR})}{I_1(\lambda_{IR})}) \quad \textbf{(Equation 15)}$$

Comparing Equation 11 with Equations 14 and 15, the Ratio of Ratios can be written in terms of the four parameters extracted by the signals provided by the photodetector and which are represented in Figure 6:

$$R_{OS} = \frac{\ln(\Delta T_R)}{\ln(\Delta T_{IR})} = \frac{\ln(\frac{I_2(\lambda_R)}{I_1(\lambda_R)})}{\ln(\frac{I_2(\lambda_{IR})}{I_1(\lambda_{IR})})} \quad \textbf{(Equation 16)}$$

Then, empirically derived calibration curves are used to determine the oxygen saturation based o R_{os}.

Calibration

The processing module receives the pulse wave, integrates and analyses the data and calculate the R_{os}. But this ratio is just an empirical measurement and doesn't provide any accurate value of oxygen saturation. So it is necessary to elaborate a calibration algorithm, to provide accurate readings [9].

In practice, a clinical empirical formula for the S_PO_2 is used [11]:

$$S = a - b\,R \quad \textbf{(Equation 17)}$$

where a and b are coefficients that are determined when the pulse oximeter is being calibrated, S is the variable S_PO_2 and R the variable R_{os} [11].

To get the values of ***a*** and ***b*** of Equation 17 that relate S_PO_2 with R_{os}, it is necessary a volunteer data performed by a CO-oximeter or a second calibrated pulse oximeter [12]. Then it is done some experiments using different samples, which different values of S_PO_2, and the value of R_{os} is collected to each one.

Coefficients a and b can be determined by performing the linear fit of the R values using the least squares method with the Equations 18 and 19 [11]:

$$a = \frac{\sum_{i=1}^{n} S_i \sum_{i=1}^{n} R_i^2 - \sum_{i=1}^{n} R_i \sum_{i=1}^{n} R_i S_i}{n\sum_{i=1}^{n} R_i^2 - (\sum_{i=1}^{n} R_i)^2} \qquad \textbf{(Equation 18)}$$

$$b = \frac{n\sum_{i=1}^{n} R_i S_i - \sum_{i=1}^{n} R_i \sum_{i=1}^{n} S_i}{n\sum_{i=1}^{n} R_i^2 - (\sum_{i=1}^{n} R_i)^2} \qquad \textbf{(Equation 19)}$$

where S_i is the S_PO_2 value measured by the S_PO_2 CO-oximeter or a calibrated pulse oximeter; R_i is the measured ratio R_{OS} that corresponds to S_i; and n is the number of measurements.

Now, a R_{os} to S_PO_2 curve is available, as the one represented in Figure 7.

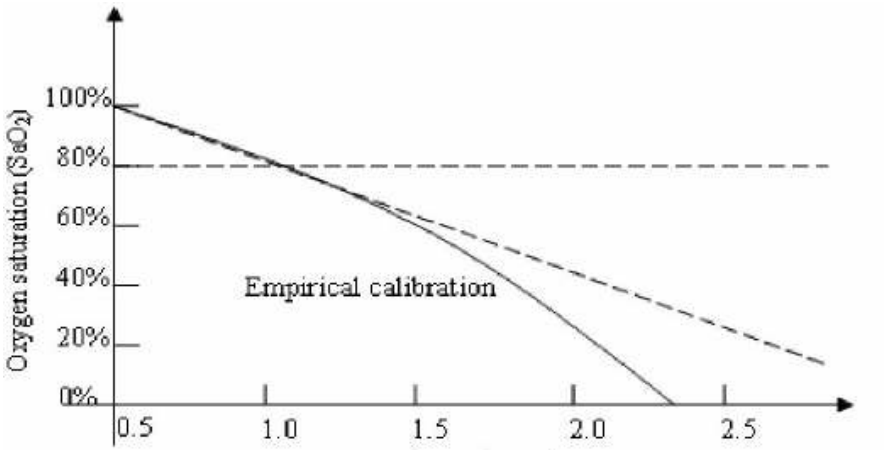


Figure 7. *Empirical R_{os} to S_PO_2 curve.* From [9]

The curve of Figure 7 shows that the S_pO_2 decreases with the increasing of R_{os}.

So using the R to SaO_2 relationship, it is possible to calculate the oxygen saturation.

2.3. Pulse Oximeters – Principles of Operation

In this section, it will be illustrate the general design and operation of a pulse oximeter. Oxygen saturation is determined by monitoring pulsations at two wavelengths and then comparing the absorption spectra of oxygenated hemoglobin and deoxygenated hemoglobin.

The biological signals have low amplitude so they can be difficult to process using common circuits. This requires that medical equipment has a special construction in order to avoid external noise and other interferences and need to operate with maximum possible safety. Schematically, a pulse oximeter can be represented by the following block diagram:

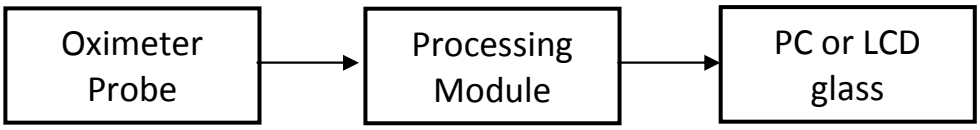


Figure 8. *Schematic block diagram of a pulse oximeter*. Adapted from [9]

An oximeter probe (Figure 9) uses two different light emitting diodes (LEDs) and each one is turned and measured alternately. The light shine through a reasonably translucent site with good blow flow. Typical adult-pediatric sites are the finger, toe, pinna or lobe of the ear. Infant sites are the foot or palm of the hand and the big toe or thumb[9].

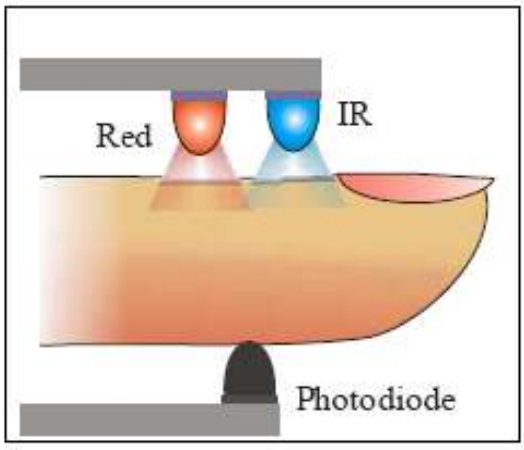


Figure 9. *The basic components of a pulse oximeter transmission probe: two LEDs with different wavelengths as light sources and a photodiode as a detector.* From [7]

There are two methods of sending light thought the measuring site: transmission method and reflectance method. The transmission method is the most common type used on pulse oximetry applications and the method used on this project; so for this discussion the transmission method will be implied. The reflectance method will be explained on the next section of this chapter.

After the transmitted red and infrared signals pass thought the measuring site, a receiver on the opposite site of the LEDs detects the output light at each wavelength. Generally, the photodetector used is a photodiode and when the light attenuated by body tissue is detected by it, the photodetector will generate a very low level current. To amplify the low amplitude signal generated and convert the current to a significant voltage, it must be used an amplifier. The most common types of amplifiers used in pulse oximetry applications today are transimpedance amplifiers which operate as current-to-voltage converters [1], [3]. Figure 10 shows the standard transimpedance amplifier configuration with a photodiode.

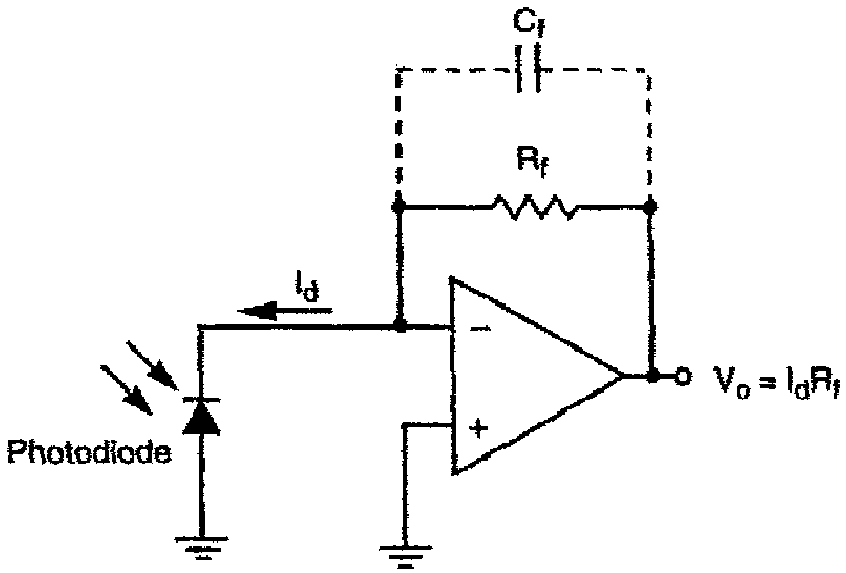


Figure 10. *Typical transimpedance amplifier used with a photodiode, where C_F is the feedback capacitor and R_F the feedback resistor.* From [1]

In this configuration, an input current is converted to an output voltage. Because of the virtual ground, the op-amp maintains zero voltage across the photodiode. Current flows through the feedback resistor and creates a voltage at the output that is proportional to the light intensity as given by [1]:

$$V_0 = I_d R_f \qquad \textbf{(Equation 20)}$$

Then the signal is sent to a processing module. This module receives the pulse wave of each wavelength, integrates and analyses the data and calculates the R/IR ratio and the heart rate. Then through an elaborate calibration algorithm based on human volunteer data it is possible to convert the ratio to pulse oxygen saturation value. Usually, the pulse oximeters use data from CO-oximeters to empirically look up a value for S_pO_2.

Finally, the results can be displayed on a LCD or transferred to PC.

The pulse oximetry method detects the changes in light absorbance through the tissues, which corresponds to the blood pulses. At each heart beat the heart contracts and there is a surge of arterial blood, which momentarily increases the blood volume across the measuring site. This results in more light absorption during the surge so light signals received at the photodetector are looked as a waveform (peaks with each heartbeat and troughs between heartbeats) [1], [6].

In the measurements of the light attenuated in the tissues, it is possible to find a direct current (DC) component and an alternate current (AC) one (Figure 11). It is assumed that the DC component is the result of the absorption by the body tissue and veins and the AC component is the result of the absorption by the arteries. The pulsatile effect occurs only in the arteries and arterioles but not in the veins. By tracking this peak-to-peak AC component, the absorbance due to venous blood or tissue does not have any effect on the measurement and only the part of signal that is directly related with the arterial blood is used for the calculation of oxygen saturation [1].

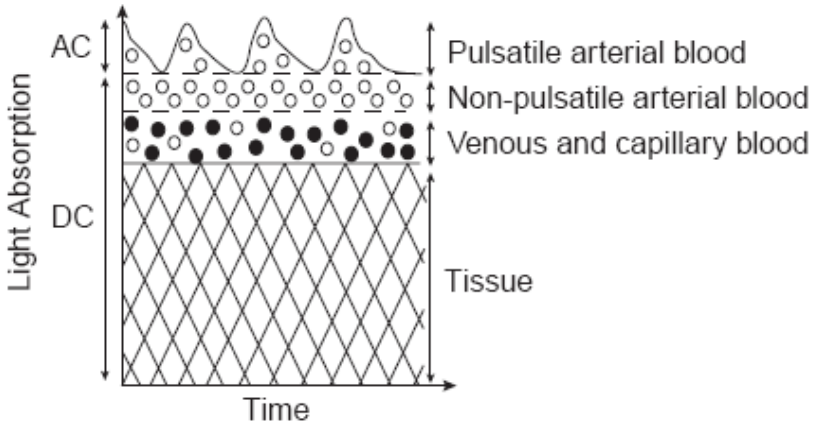


Figure 11. *Schematic representation of light absorption in adequately perfused tissue.* From [6]

Figures 12(a) and 12(b) show typical pulsatile signals detected when red or infrared LED is shone through a finger.

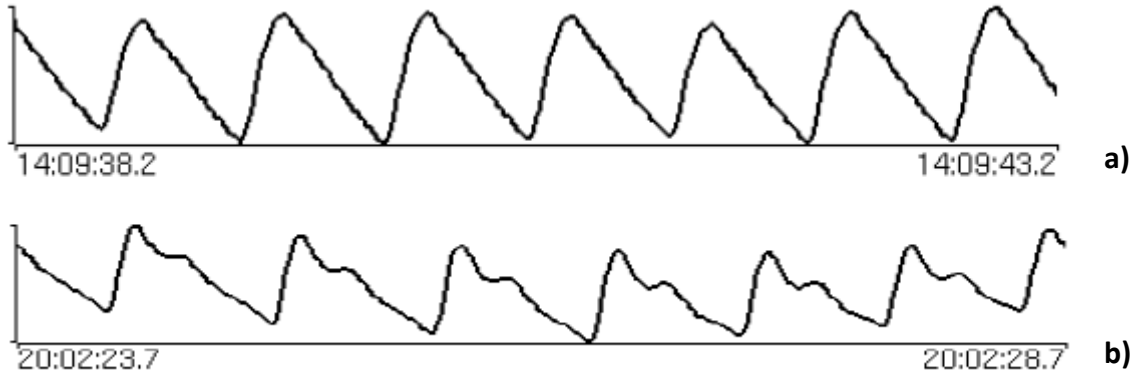


Figure 12 (a) and (b). *Typical pulsatile signals detected in the intensity of light when light passes through a finger.* From [12]

In this figures it is possible to see that the baseline (DC content) has been removed from these curves. It is also possible to see that there is quite a wide variation in the shape of the curves between different people. Particularly, in the signal of Figure 12(b) it is possible to identify a secondary peak for each heart beat, which is known as the dicrotic notch. This notch is a quite common physiological phenomenon and is a result of the sudden closure of the aortic valve which causes a momentary elevated rebound in the pressure reading, such that the smooth downward slope of the pressure waveform is interrupted by a very brief upward movement forming a sawtooth notch [12], [38].

2.4. State of the Art

The techniques of oximetry and particularly the non-invasive methods play an important role on healthcare field. The method of pulse oximetry is the most used today but it is possible to find alternative optical methods which compete with it. In this section, it will be described these existing methods.

It will be exposed also an analysis of the features of the pulse oximeters that can be found in the market and especially what exists available for oximeter probes.

2.4.1. Alternative Non-Invasive Optical Methods of Oximetry

- **Ear Oximetry**

The first non-invasive oximeters that appeared were the ear oximeters, around 1935 [6], when it was proved that the transmission oximetry could be applied to the external ear. In this oximetry, the light of one or more wavelength is transmitted through the ear lobe or the pinna of the ear of the patient and the intensity of transmitted light is measured on the other side of the ear lobe [10].

However, the major inconvenient of ear oximetry was revealed to be the inability to differentiate light absorption due to arterial blood from that due to venous blood and tissues. In order to overcome this, the newer devices also make arterialization of the blood capillaries to dilate the vascular bed in the area of measurement and thus increase the infusion. Moreover, the devices compare the optical properties of a "bloodless" earlobe (by compressing it using a special device) with the optical properties using a perfused earlobe [6], [10].

An example of an ear oximeter is manufactured by Hewlett-Packard that developed the model 47201 ear oximeter (Figure 13). This oximeter is based on light transmission at eight wavelengths, using a high intensity tungsten lamp that generates a broad spectrum of light wavelenghts. This light passes through filters of light before entering the fiberoptic cable, which carries the light to the ear. A second fiberoptic cable carries the transmitted light pulses of the ear to the device for detection and analysis [6], [12].

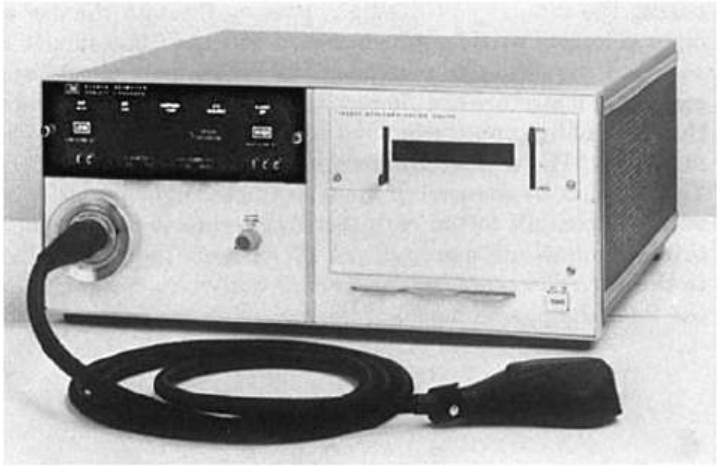


***Figure 13**. The Hewlett Packard Model 47201A ear oximeter.* From [6]

The probe used in the ear is relatively large (10x10cm) and is equipped with temperature-controlled heater, to keep the temperature at 41 ºC (ear must be at this temperature for capillary blood arterialization), thus presenting a discomfort associated with the warming of the ear [6], [10].

Due to the volume of the components involved, the high cost of instrumentation, the need for measurements at eight different wavelengths and the development of technology to solve many of these problems, the ear oximeters is no longer commercialized [6], [9].

- **Laser Oximetry** [14]

Laser oximetry is a new non-invasive method to evaluate changes in tissue perfusion and determine the oxygen saturation on targeted areas of tissue, through a continuous wave optical spectrometer operating in the near-infrared spectrum.

Nowadays, there are small oximeter probes for measure the oxygen saturation on extremities of human body such as finger or earlobe through the transmission of light through a vascular bed to a detector in the opposite site of the emitter. Thus, there are limitations on the use of these probes for larger organs or tissue structure.

Recently, the development of fiber optic array probes including multiple light sources and one photodetector coupled to a continuous wave optical spectrometer makes possible to penetrate a targeted volume of tissue (500 mm^3) to an average depth of 5–8mm beneath the skin surface. The photodetector fiber is placed on the same side of the source fibers and collects scattered light.

One of the current applications of laser oximetry is the measure of blood flow in the fetal brain, using multiple source probes. However, it is a very recently method and its viability for various parts of the body is not established.

2.4.2. Pulse Oximeters and Oximeter Probes

As said previously, by taking advantage of the pulsatile component of blood, the pulse oximetry is able to overcome many of the problems of earlier technologies. Pulse oximetry is a technique that has been for a long time developed and improved, in order to decrease its limitations, which would lead to better patient care.

Despite the newer technologies, pulse oximeters still present some problems such as the accuracy of the measures, which has been shown to be ≈ ±4% when compared to arterial blood oximetry measurements (S_aO_2) [6]. As it was said previously, the presence of abnormal hemoglobins such as carboxyhemoglobin and methemoglobin can lead to some erroneous readings affecting the accuracy of the device. When the presence of either of these hemoglobins is suspected, pulse oximetry should be supplemented by in-vitro multiwavelength CO-oximetry. Another limitation is the response in time because there is a delay between a change in S_PO_2 and the display of it [8].

In the market, there are different pulse oximeters worked out by many manufacturers, that offer solutions with many differentiating factors between products. However, there are common features of the pulse oximeters that can be found in the market [40], [41]:

- the devices are specially designed to measure arterial oxygen saturation but most of them can also measure the heart rate and the plethysmography wave;
- they are useable in children and adults;
- the most part have small dimensions to allow a better portability and less discomfort to the patient;
- they are wearable and non-invasive;
- most of them have very low power consumption.

There are pulse oximeters that provide a continuous monitoring of the vital parameters and other ones that provide a discrete monitoring of the same parameters. The first ones are power sourced by regular batteries with an equivalent use range of 20 hours and able to record the data equivalent to that period, useable under a relatively broad temperature and humidity conditions range for general purposes. These ones give the results with 2/3 digits precision. The second devices are also power sourced by regular batteries with autonomy for approximately 20 hours, two digits precision for oxygen saturation and are usable under a relatively broad temperature and humidity conditions [40], [41].

The market analysis becomes clear that there are pulse oximeters whose oximeter probe and the processing module form just a single device, which provide a better portability.

Figure 14 shows a modern pulse oximeter designed by Nonin, which incorporates the electronics and sensor into one single unit. This device provides information about SpO_2 and pulse rate which can be read from any angle. It operates on two alkaline batteries for approximately 1600 spot-checks or up to 18 hours of continuous use and it accommodates a wide range of finger thicknesses [15].

Figure 14. *Portable Nonin Onyx 9500 pulse oximeter.* From [15]

There are also oximeters (which is what often happens) that are integrated with more complex systems and can be connected with a variety of oximeter probes. That is the case of the oximetry solution of the project and this thesis particularly focuses the

development of an oximeter probe so it makes sense to include on the discussion the different types of oximeter probes which can be found in the market.

There is a specific market to the oximeter probes, which can be commercialized separately and then connected to an acquisition and processing modules. As said previously, the oximeter probes are used to acquire the biological signal and send the data to measure the oxygen saturation in blood. A flexible cable connects the probe to the pulse oximeter unit, carries electric power to the LEDs and the signal from the photodiode.

Generally, oximeter probes present some limitations such as the detection of arterial pulse in low perfusion states (hypotension, hypothermia, cardiopulmonary bypass or low cardiac output), where can difficult to distinguish the light absorbed by arterial blood and tissue from that absorbed by the venous blood and tissue [6], [8]. Brown, blue and green polish nails may affect the calculation of S_PO_2 too. Other problems of oximeter probes are the motion artifacts that affect the results of oxygen saturation and the ambient light interference.

There are two main types of probes on oximetry: transmission probes and reflectance probes. The difference is in the position of the photodetector on the probe as is shown in Figure 15.

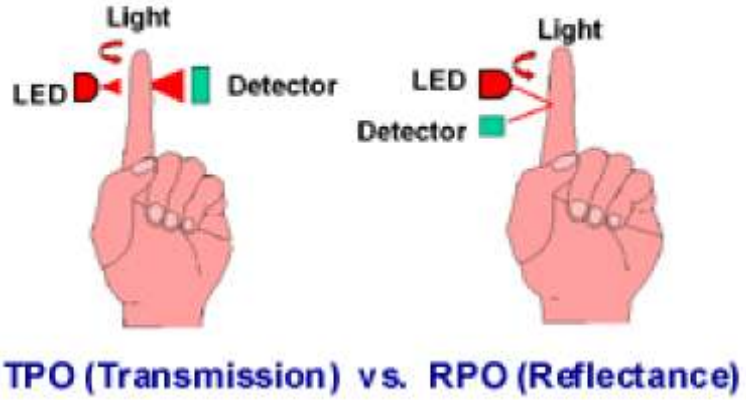


Figure 15. *Transmission vs reflectance oximeter probes and the position of their components.* From [16]

The transmission probe has two LEDs on one side and the photodetector on the other and the site of measurement is inserted between the two. So a pulse oximeter with transmission probes uses the light transmitted through an extremity to measure the blood oxygen saturation. Figure 16 shows a general example of a transmission probe.

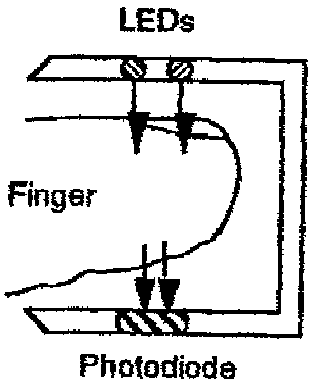


Figure 16. *Transmission Probe: light emitted by the LEDs passes alternately through an extremity of the body and the transmitted light is detected by a photodetector (a photodiode in the figure).* From [1]

The LEDs of those probes are powered alternately; the light of each wavelength will pass through the tissue and the photodetector will detect the transmitted light, which was attenuated by the amount of blood present in the tissue. As the amount of blood varies with the arterial pulse, transmission probes are used to give also information about the heart rate.

The light sources and the photodetector are placed facing each other so that the maximum amount of light can be detected. The photodiode also is positioned as close as possible to the skin without exerting force on the tissue [1].

On the other hand, a reflectance probe has the LEDs and the photodetector on the same side. It must be placed over a point with underlying bone. Figure 17 shows a general example of a reflectance probe.

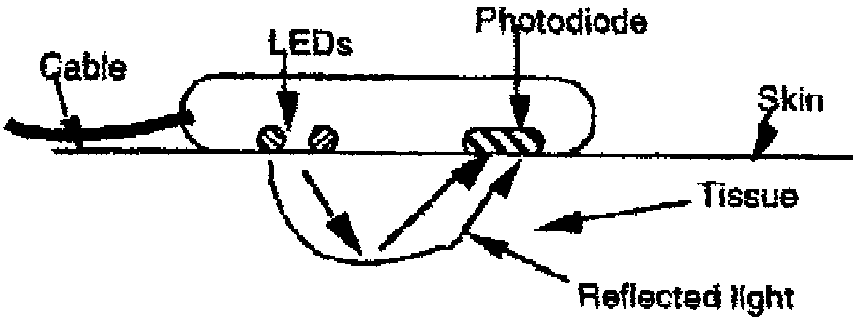


Figure 17. *Reflectance Probe: light emitted by the LEDs enters in the tissue, is scattered and is detected at the photodetector (a photodiode in the figure).* From [1]

The LEDs are powered alternately, the light passes through tissue and blood vessels, is scattered by moving red blood cells and nonmoving tissue and a part of this back scattered light passes through the tissues again and is then detected by the photodetector. The output signal is processed by the pulse oximeter unit, which measures the S_PO_2 of the blood.

Comparing the both oximeter probes, it is important to refer that a significant amount of light will reflect off the skin in the reflectance probes and this light will be detected, which doesn't occur in the transmission probes. Moreover, reflectance probes have a high offset and a lower signal-to-noise ratio when compared with the transmission probes. However, reflectance probes require a significantly greater amount of light so either more LEDs or more photodiodes need to be used [1].

Within the oximeter probes mentioned above, there are several types used for different parts of the body. Transmission and reflectance probes are used clinically, though transmission probes are more common due to the simplicity of signal analysis and the ease of attachment and remove.

Due to its principle of operation, reflectance probes can be virtually used in any place on the human body where the probe can be placed while the transmission probes can only be used on limited parts.

Transmission probes are more commonly placed on the finger or earlobe. Reflectance probes can be used at multiple sites of the body, not necessarily extremities,

such as forehead, temple or sternum. On the next paragraphs, it will be described the main types of reflectance and transmission probes that can be found on the market.

The finger transmission probe consists of a spring-loaded clamp which attaches to the finger, as it is shown in Figure 18. One arm of the spring-loaded clip contains the two LEDs and the other contains its photodetector, thus allowing the probe to measure the light transmitted through the finger.

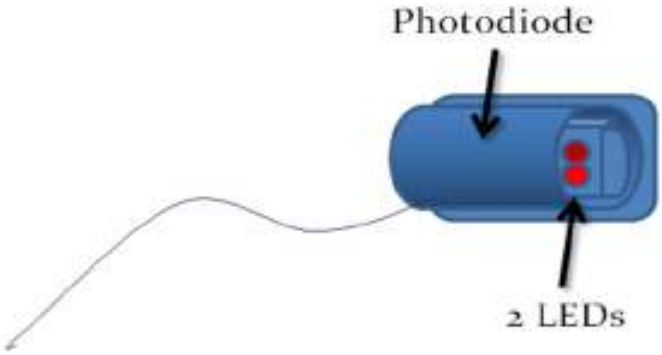


Figure 18. *Finger Transmission Probe.* From [16]

The main advantage of those probes is its ease to quickly apply it to a finger of a patient and quickly remove it. Moreover, they do not present discomfort to the patient and are unobtrusive. However, these probes have some limitations such as [16]:

- they present a low level of mechanical resistance, especially after multiple uses will be under great mechanical stress which means that the probes won't be very durable;
- difficulty in designing a probe that can be used both for infants and adults finger, so it depends on the patient morphology;
- physiological conditions such as a situation of low blood pressure after a loss of blood can affect the accuracy of readings since the body reduces the blood flow to the periphery to maintain adequate blood pressure for the vital organs and brain.

The earlobe transmission probe consists of two prongs which would be placed on opposite sides of the earlobe: the LEDs are placed on one prong and the photodetector on the other one, as shown in Figure 19.

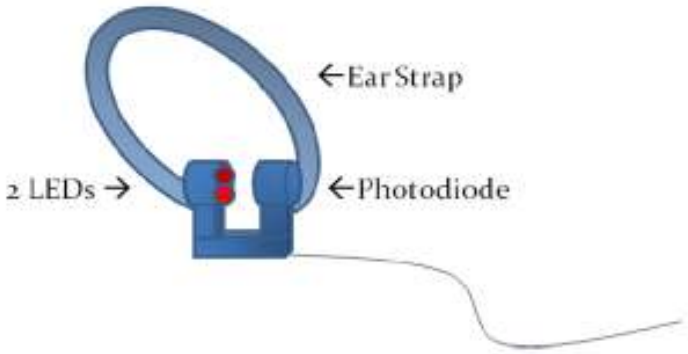


Figure 19. *Earlobe Transmission Probe.* From [16]

Unlike the transmission finger probe, the earlobe probe uses the reliable perfusion of the head which remains perfused even in cases of severe shock, in which the patient's peripheral circulation may be cut off. So those probes are more suitable for patients with severe blood loss. Those probes have also the advantage of being relatively easy to use, requiring little or no adjustment between patients. However, the main disadvantage of the earlobe probe is that it may need to be adjustable to be used on both adults and small infants [16].

Within the reflectance probes the more common type is the forehead reflectance probe. This consist on face of the disc that has two LEDs in its center, surrounded by a ring of three or four photodiodes, as shown in Figure 20.

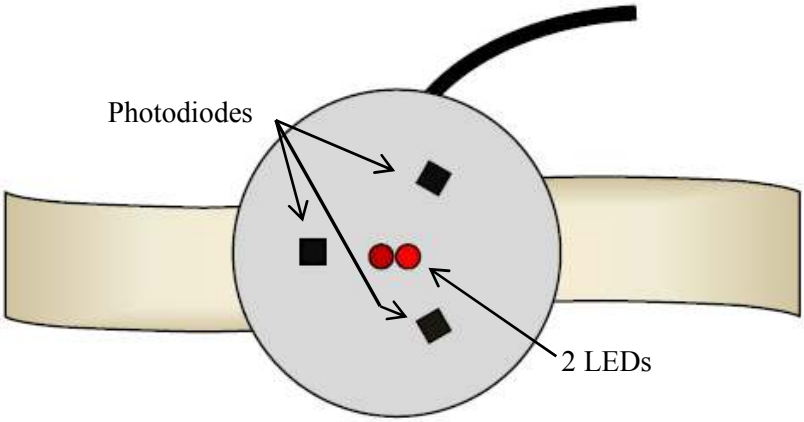


Figure 20. *Forehead Reflectance Probe.* Adapted [16]

These probes have no moving parts and present a better level of mechanical resistance, being probably more durable. Moreover, they have the advantage that is common in the reflectance probes: the fact of being used on both adults and infants

without problem, as they just require a relatively flat perfused tissue backed by bone. Like earlobe transmission probe, forehead uses the reliable perfusion of the head [16]. However, the signal resolution of reflectance probes is lower, creating a higher possibility of an inaccurate reading. Also, the probe is not well suited for spot checks; the headband would need to be adjusted frequently between patients. Some physicians might manually apply the probe, which could potentially result in motion artifacts. Also, with constant readjustment, the headband would need to be periodically replaced [16].

Finally, in the market it can be found reusable and disposable probes. The reusable probes are all probes with nonadhesive or disposable adhesive sensors and their main advantage is obviously its low cost involved. However, reusable probes involves some drawbacks including the inconvenient of require cleaning between different patients to minimize the risk of cross contamination and the fact that they are more susceptible to signal distorting motion artifacts [1].

Disposable probes are the ones that are discarded after they have been used, eliminating the risk of cross contamination between patients. Generally, the disposable probes are adhesives so they decrease the signal distorting motion artifacts, because it is possible to secure the probe in the proper position [1].

2.5. Test Procedures (ISO 9919:2005)

ISO 9919:2005 is a document that presents the requirements for the basic safety and essential performance of pulse oximeter equipment intended for use on humans. These requirements also apply to pulse oximeter equipment, including pulse oximeter monitors, pulse oximeter probes and probe cable extenders that has been reprocessed.

In the Attachment A, it is possible to find the first eleven pages of this document (the remaining part should be paid to have access). ISO 9919:2005 provides a wide range of information that represents a guideline in the manufacturing process, such as the classification of the device, the technical description, components and its assemble, calibration, the requirements for tests or accuracy of operating data.

2.6. Commercialization of Pulse Oximeters

Once the device is completed and all the tests have been passed the following step is to register the product.

On the beginning of this project, it was proposed to the students the study of the necessary procedures to market a medical device such as a pulse oximeter. On this section, it will be presented the research done with the notified organisms.

For medical devices (except for custom-made and intended for clinical research) to be available for commercial sale, they must meet the requirements of specific legislation. In Europe if devices meet the requirements, they will have a verifying marking on the outside, CE Marking. In US, Food and Drugs Administration (FDA) is submitting the certificate (510k). These markings have a very specific graphical look and should be placed by the manufacturer in a legible, visible and indelible way.

- **CE Marking**

Figure 21. *The CE marking design.* From [17]

The CE marking is like a mark of product quality and it is a declaration that the product meets all of the appropriate provisions of the relevant legislation required to implement specific European Directives. Focusing in pulse oximeters, they should respect Medical Devices Directive 2007/47/EC [17].

The approved medical devices present the CE Marking and in addition a code of four digits which is the identification number of the notified body chosen by the manufacturer for evaluation. This notified body is responsible to check if the device is

according with the requirements and carries out the procedures for conformity assessment. Finally, in case of assent, the notified body issuing the CE certificate of conformity which would allow the manufacturer to affix the CE marking on its medical devices [17], [18].

The entity which regulates the certification of medical devices in Portugal is INFARMED - Portuguese regulatory authority that evaluates, authorizes, regulates and controls medical devices, according the procedures in Directive 2007/47/EC [18].

In order to initiate the medical device certification process, the manufacturer should compile a file with the necessary documentation: an application, a statement of commitment that the manufacturer did not request the evaluation to another notified body and all the necessary scientific technical documentation in accordance with the procedure chosen from those provided by Directive 93/42/EEC (the device specifications, technical tests to the device, the construction techniques employed and a technical description) [14]. The documentation to fill and the information about the organization of the dossier is possible to find in the Infarmed web page [18].

- **FDA(510k)**

FDA(510k) is a certificate that allows the marketing of products and medical equipment in the US. The agency responsible for this issue is the Food and Drug Administration (FDA, US), which is dependent on the Government's health area. Generally, manufacturers/importers of some Class 1 and Class 3, and most Class 2 medical devices, are required to file a 510k. So it is important to determine the classification of the new device using the database online at FDA site. To FDA an oximeter is a Class 2 medical device and belongs to the cardiovascular monitoring devices [19].

A 510(k) requires demonstration of substantial equivalence to another legally US marketed device which means that the new device is at least as safe and effective as the predicate. A device is substantially equivalent if, in comparison to a predicate it has the same intended use and has the same technological characteristics as the predicate or has different technological characteristics and the information submitted to FDA demonstrates that the device is at least as safe and effective as the legally marketed device [19], [20].

In order to regulate the medical device it's necessary to be subject to a number of steps. First it is necessary to compile the information needed and prepare the submission, thought the organization of a file which includes a list of specific information that will be required in the application, such as an executive summary, an intended use and a technical description of the device [20].

FDA does not have a template for 510(k) submission, so the company needs to figure out how to meet their requirements in submitting all of this information, which is proving not always easily accomplished.

Once the company has submitted the file, the FDA reviews the 510(k) application, which can last up to 90 days. During this period they may ask for additional information at which time the "clock" is stopped and then resumed upon the FDA's receipt of the answer to their questions. If approved, the FDA will send a letter, with an assigned 510(k) number, that says they "*have determined that your device is substantially equivalent to legally marketed predicate devices...and you may therefore market the device subject to general controls provisions of the (Food, Drug and Cosmetics) Act*" [20]. This letter means that the device is much the same as the predicates already approved by the FDA. The letter will be available on the FDA database as proof to the future customers that the device is approved for sale in the US. This order "clears" the device for commercial distribution.

Finally, once the company has received the FDA 510(k) "clearance" letter it is necessary to complete the FDA device listing and establishment registration using a system in the FDA website and then the company and the device are registered with the FDA. There are also certain fees that must be paid [20].

As said previously, the FDA does not provide a 510(k) template to follow and that is why many people find them very difficult to complete properly. This is also why it is very difficult to reach to a fixed price of how much costs to prepare and submit your 510(k) application to the US Food and Drug Administration.

In the Attachment B, it is possible to find information of a consulting group called EMERGO, that mediate the process and, because they have completed many applications, they are able to offer a fixed price to prepare and submit a 510(k) application to the US Food and Drug Administration.

3. ARCHITECTURE OF THE SYSTEM

3.1. Project Requirements

In the logic of the work and under the guidelines of the project suggested by ISA and GEI, the students intend to develop a finger transmission oximeter probe. This probe should be simple, reliable, robust, economic and ensure the correct switching between red and infrared channels through a simple analog control signal (square wave), and ensure also the correct detection of the transmitted light spectra for both producing a consistent signal amplified. The sampling frequency (repetition rate at which the red and infrared LEDs drivers) should be well above the maximum frequency present in the arterial pulse (around few Hz). Keeping in mind the future portability of the device, the circuits should be powered at 5V because the commercial batteries are around this voltage.

The oximeter probe should interface with a pulse oximeter portable unit, which includes an acquisition and processing module. The acquisition module must have an acquisition rate that is at least twice the sampling frequency, according to the sampling theorem so that the events could be detected. The processing module must have a processing and memory capacity that allows the implementation of algorithms for determining the heart rate and oxygen saturation in the blood. This module should also have accessories processes which provide some noise reduction, avoid erroneous readings. The processing module must return results in a discrete way. As the oximeter probe, this modules should be powered at 5V due to the portability and ensure the least possible power consumption.

3.2. Work Evolution

On the beginning of the project and after the study of the Theoretical and Technical Background, the students developed an oximeter probe in order to start reading signals since the main objective of the project was the development of a portable pulse oximeter unit. That oximeter probe was assembled based on the circuit presented in the Attachment C.

The circuit includes SMD components: an infrared LED, a red LED, two polarization resistors, a decoupling capacitor, a photodiode, a transimpedance amplifier, a feedback resistor, a feedback capacitor and two other resistors in the voltage divider. It is also used a Darlington driver which has the function of control the current that passes to the LEDs. So, in unusual cases, where the circuit sinks too much current which can damage its components, the Darlington driver protects the circuit.

The project team did some research that allows choosing the most appropriate values for resistors and capacitors, taking into account the bandwidth required and ensure the smooth functioning of the probe.

It was chosen LEDs that operate in the range specified in the absorption curve of the Figure 4. It was used red LEDs of 635nm and infrared LEDs of 950nm. The LEDs have an internal system of lenses that provide high-brightness output, to get better results.

Several tests were done to choose the values of polarization resistors so that the light from the LEDs which passes through the finger is enough to be detected by the photodiode. Thus, these resistors have a value of 100Ω.

The decoupling capacitor is just to prevent an eventual coupling between components via the power supply connections, so its choice is not really important and it was chosen the value of 10nF as an acceptable value.

The detection of the light transmitted through the finger was done using a photodiode with a transimpedance amplifier configuration. Although it is the most frequently configuration used in pulse oximetry applications, the transimpedance configuration has a number of multidimensional constraints so important considerations must be taken into account on the choice of the components. These considerations are essential in the choice of the values of feedback resistor and feedback capacitor and a

detailed discussion of it can be found in Section 3.4.1.2. That study led to the choice of the values of 33KΩ to the feedback resistor and 10nF to the feedback capacitor.

The resistors of the voltage divider have the value of 1KΩ in order to ensure that the voltage in V+ is 2,5V and the signal is around this value.

The circuits were assembled in a black box in order to build a prototype of the oximeter probe, as shown in Figure 22 and Figure 23.

Figure 22. *First oximeter probe prototype developed*

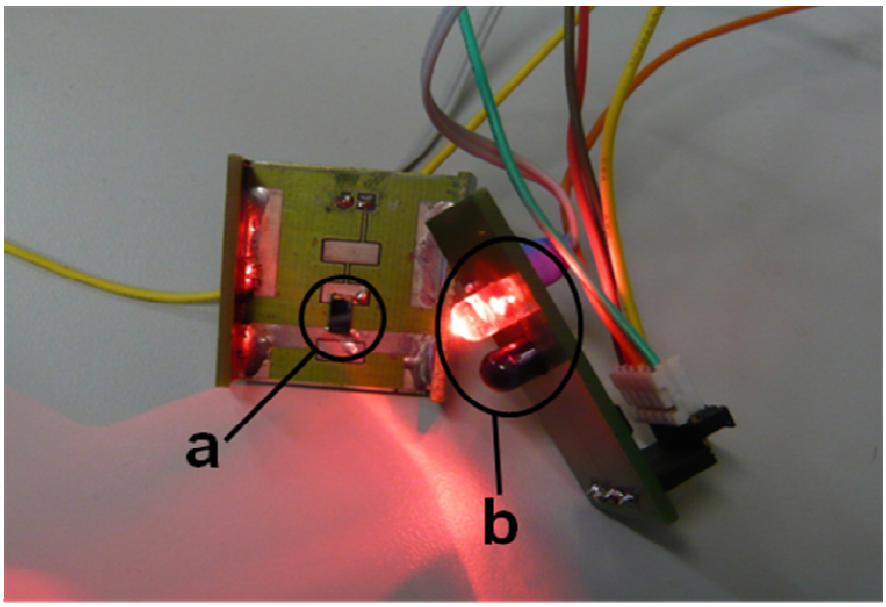


Figure 23. *Oximeter probe prototyte on; a – photodiode; b – red and infrared LEDs (red one in operation)*

Once the oximeter probe prototype was assembled, each LED was powered in each time, in a continuous way, in order to test the prototype. It was used a NI-6009 board to acquire and store the signals in a text file, with a frequency of acquisition of 300Hz, which is well above the maximum frequency present in the arterial pulse.

The students developed a program in Matlab, which allow viewing and store the signals acquired, for a further analysis with some Matlab algorithms in order to minimize the noise and to make the peak detection to determine the heart rate. The algorithm implemented in Matlab can be found in the Attachment D and it will be explained in the Section 3.5. This algorithm was used for detecting the maximum and minimum of the acquired signal, after a MATLAB simple processing to reduce the noise. This algorithm does not interfere with the treatment of the signal by the processing module; it is used only to evaluate the quality of the signal and identify the heart beat pattern.

The work described has led to some preliminary results: inserting the finger in the probe and using each LEDs alternately, are obtained the curves represented in the Figure 24. The following two signals were from two different people.

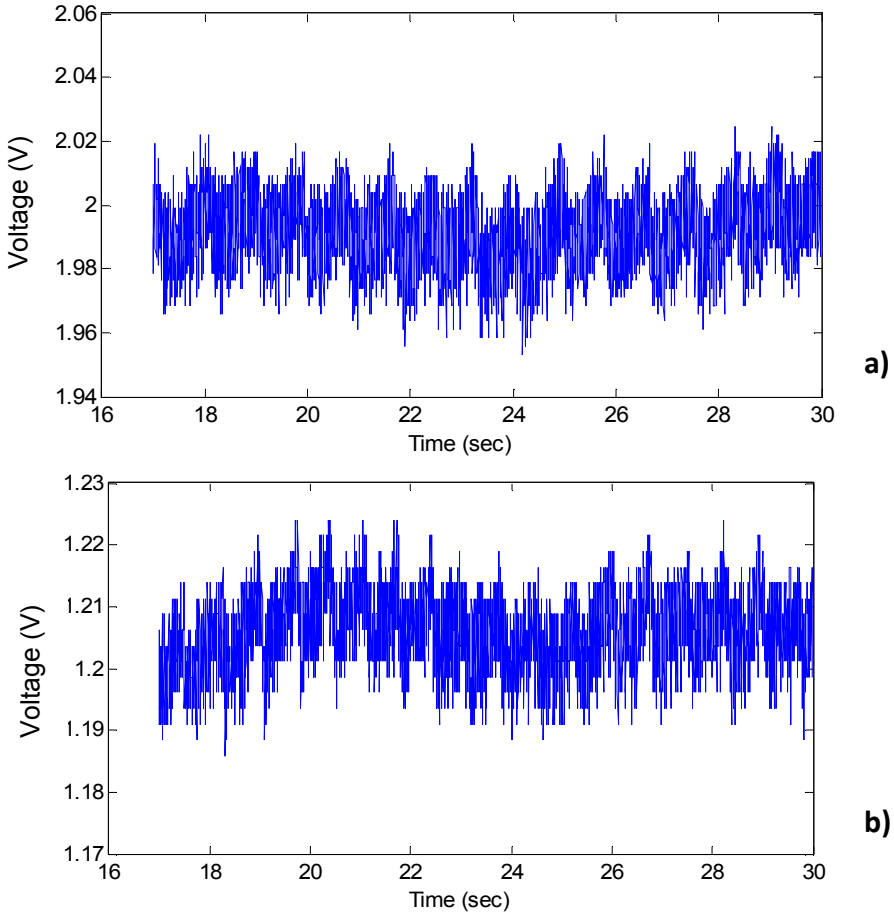


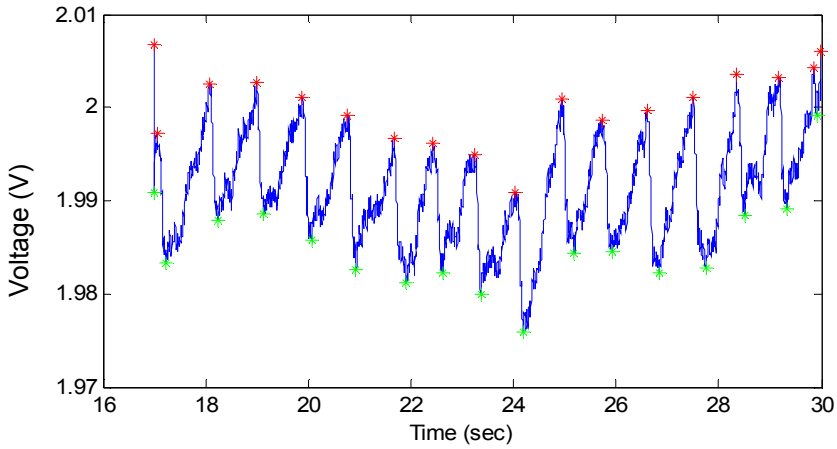
Figure 24. *graphical representation of the data acquired with the NI-6009 DAQ, between the 17 and 30 sec;* ***24a)*** *- using the infrared LED;* ***24b)*** *– using the red LED*

In Figures 24a) and 24b) it is possible to see that there is a significant noise; nevertheless, it is possible clearly to see a pulse signal. In 24a) using the infrared LED, it is getting voltages between 1,96V and 2,02V (signal is around 60mV); in 24b), using the red LED, it is getting voltages between 1,2V and 1,23V (signal is around 30mV)

However, the signal acquired presents a significant band noise which difficult the detection of maximums and minimums in the curves. In order to get a better signal, it was applied to the signals of Figure 24 the program developed on Matlab.

After the treatment by Matlab the curve in the Figure 25 was obtained; this curve is clearly better defined, being visible a pulsatile pattern, allowing the detection of the maximums and minimums to the proper pulse rate determination.

a)



b)

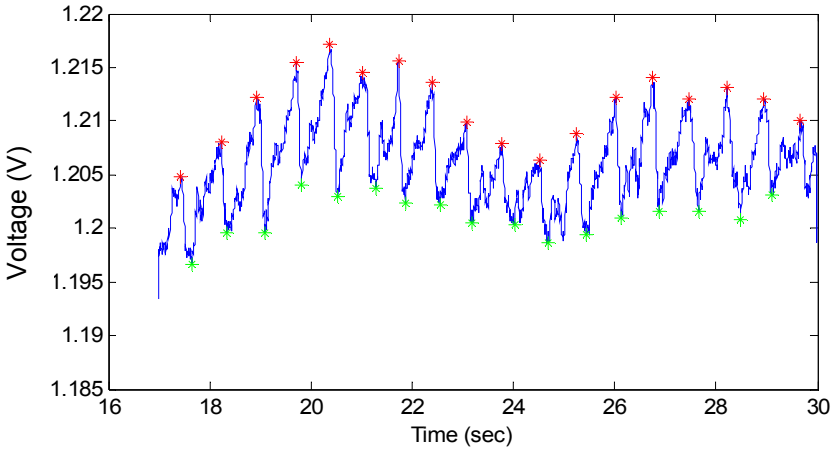


Figure 25. *graphical representation of the signal acquired with NI-6009 DAQ, between the 17 and the 30 sec after Matlab processing* ***25a)****- using the infrared LED;* ***25b)*** *– using the red LED*

In Figure 25 a) and b), it is possible to see a pulsatile signal with less noise than the previous where is possible to detect the maximums at red and the minimums at green, through the algorithm of peak detection implemented in Matlab; this detection allows the heart beats counting and consequently the heart rate. In the signal of Figure 25 a) the heart rate returned by the program is 72 BPM while the signal of Figure 25 b) the heart rate determined is 84 BPM.

However, several tests were done and the results became clear that the probe architecture was not the most convenient due the absence of consistent results and data repeatability. So it was necessary to test different solutions, the SMD platform in use was not suitable to perform different components and circuits so it was decided to implement the new probe development on a breadboard. The new work orientation coincided with the ISA wakeup to the useful interconnection of our project with their, this reinforced collaboration allowed the project team to use ISA platforms, and at the moment the stand alone device was bypassed.

3.3. Overall Architecture of the System

Regarding the new goals proposal the initial aim of a stand alone oximeter was bypassed, not overwhelmed, thanks to the knowledge available and used in other similar projects by ISA. New project requirements were described in Section 3.1. So the main objective of the project became the integration of the previous results and developments of the project in the ISA platform. Figure 26 shows the overall architecture of the system.

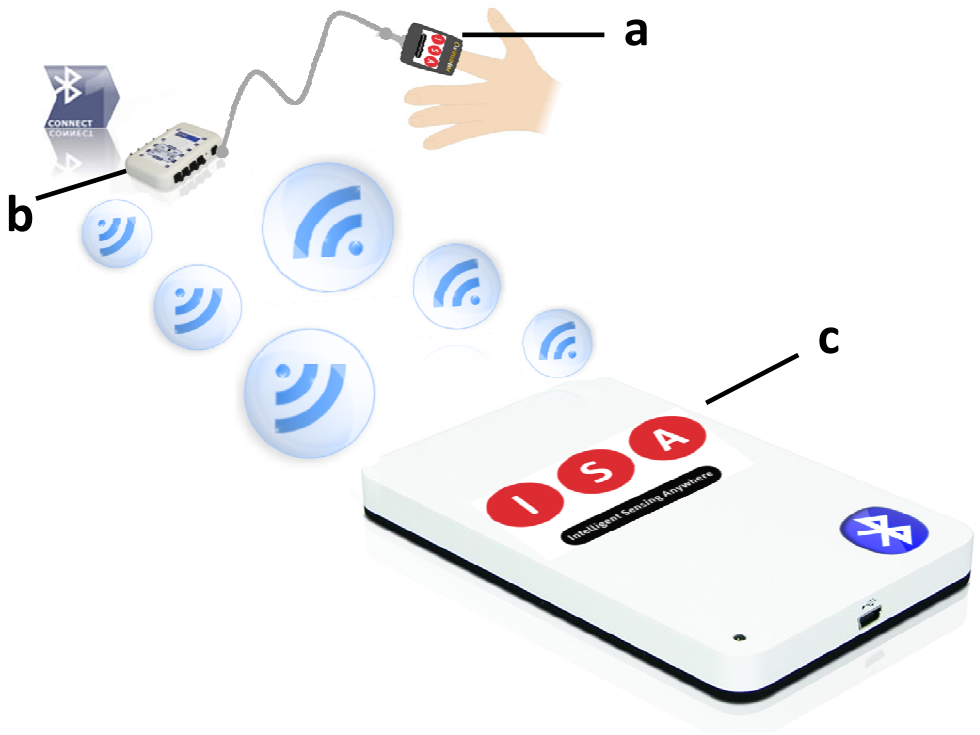


Figure 26. *System Architecture; a - oximeter probe; b – acquisition module; c – processing module*

The project is mainly divided in two functional areas: hardware part which includes the development of an oximeter probe and a software and firmware part including the development of the signal analysis processes. Ana Domingues was responsible for the hardware development of acquisition modules. The partner of the project [39] was responsible by the development of algorithms in the microcontroller that allow the signal processes and the determination of the heart rate and the Ratio of Ratios which will be used to the calibration of the pulse oximeter.

3.3.1. Oximeter Probe

The oximeter probe should be integrated on the Bluetooth™ module in the BioPlux™ that provides the signal control and ensure the analog/digital conversion of the signal produced by the probe; the signal will be transmitted via the Bluetooth™ protocol in real time for a concentrator module responsible for processing it.

To develop the oximeter probe, the first step of the work is to choose the wavelength that will be used, according the absorption of light by the hemoglobin. Once it was done, it should be projected and assembled the three different modules that constitute the oximeter probe: emission module, reception module and timing module.

The LED Driver Module consists of two circuits of polarization, one for the red LED and one for the infrared LED in order to switch them. This circuit is powered by a voltage of 5V and integrates various economic electronic components such as diodes, transistors and resistors. This module should ensure that the current which passes through the LEDs will be constant to ensure a constant optical power and the circuits should be projected so that the red and infrared LEDs can operate alternately.

The Photodetection Module is constituted by a photodiode that detects the light that was transmitted through the finger and convert the current to voltage by a conversion circuit of current-voltage, using a typical transipedance amplifier. The gain of the circuit is given by the value of feedback resistor and it is placed in parallel a capacitor to reduce high frequency noise and the possibility of oscillations.

The acquisition module (BioPlux™) has only one digital output port so it is not possible to multiplex the LEDs through it. Therefore, it was necessary to develop an electronic circuit that could be connected to the driver circuit of the LEDs and proceed to the timing of them. The timing circuit was built around the 555 timer and the NAND gate integrated circuits.

3.3.2. Acquisition and Processing Module [39]

The ISA platform is formed by the *Leonardo*® board synchronized with a BioPlux™ module (by Bluetooth™ connection with 100m range). The BioPlux™ module has 6 analog channels with 12bits and a sample rate of 1000Hz and 2 analog channels with 12bits and a sample rate of 125Hz available to connect to the oximeter probe and perform a signal digitalization. It has also 1 Digital I/O Port with 1 bit. All the firmware to board initialization (configuration bits for CPU and bus speed (Hz), timers, interrupt service routine) and communications protocols (I^2C, SPI, UART, Bluetooth™) was already developed and are operational so the task was to continue developing the signal analysis related processes: signal linearization and smoothness (with a derivative, squared and moving average), maximums and minimums (peak and valley) detection to calculate the AC and DC component for the different channel (R and IR) and use that values to calculate the R_{OS} and determine the heart rate; all this algorithms have to be implemented in C language.

Regarding the signal processing module in the concentrator module, the algorithm should first clear the noise of the original signal (moving average, junction, potentiation, threshold) and subsequently identify the peaks in the signal for each of the spectra, store the values in different vectors and use them for implementation the algorithm to determine the oxygen saturation. The oxygen saturation is obtained by a simple association between the ratio of peak values in two consecutive transmission spectra and the different values of a saturation table. The preparation of a table involves creating a calibration curve as was mentioned in Section 2.2.

The algorithms should be robust to motion artifacts and this fact has some possible approaches mentioned in the literature associated with few details. However, in any case it will be associated to assigning an importance index to each reading according to the local history of acquisition; for example a portion of signal without major changes will have a greater weight than one in which there is a sudden change in the value of the signal. The concept is outlined and now needs a real interpretation and adjusted to the needs of the project. The final result will be a value of saturation, which is just an integer and the display of the value is dependent of ISA requirements.

3.4. Oximeter Probe Design

To the building of this new prototype it was used two arms that attaches to the finger using a Velcro tape, as it is shown in Figure 27. One arm is a part of a spring-loaded clip and contains the two LEDs (Figure 28 a)); the other arm is a black piece of plastic and contains the photodiode which is protected by a translucent piece (Figure 28 b)).

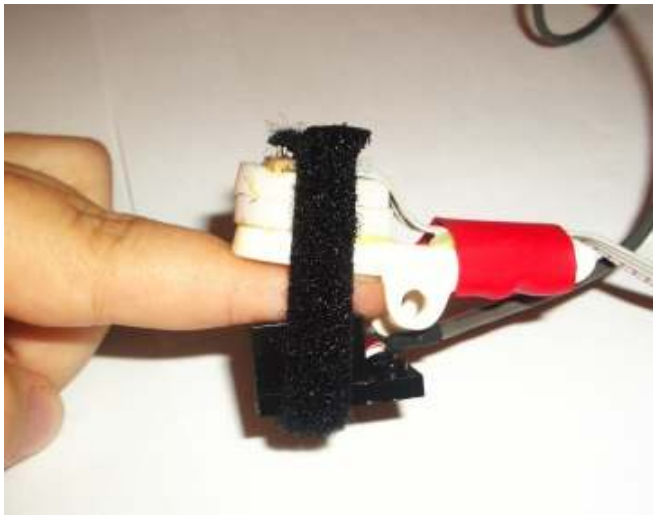


Figure 27. *Oximeter Probe Prototype; a – arm containing the two LEDs; b – arm containing the photodiode*

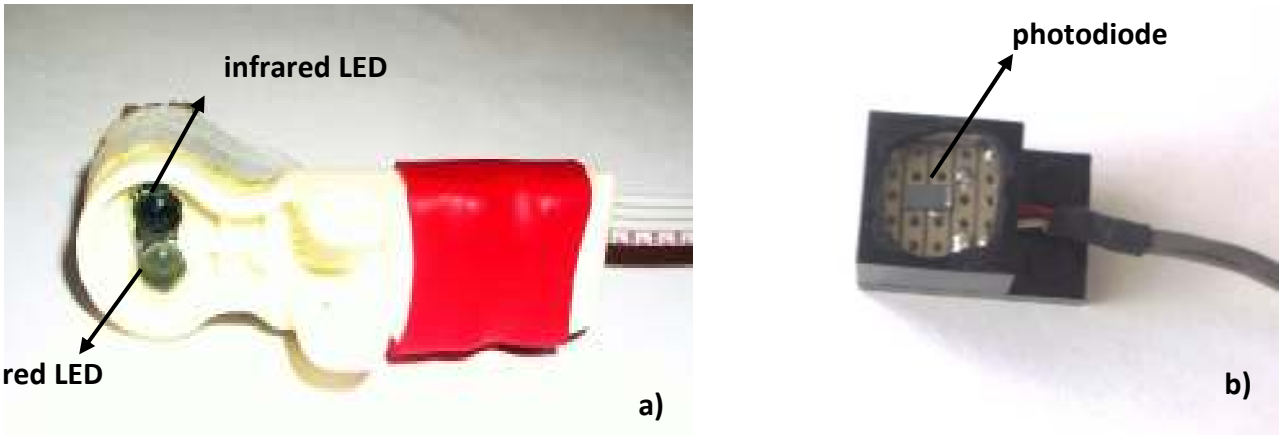


Figure 28. *Arms of the Oximeter Probe Prototype.* ***28)*** *– arm containing the red and the infrared LEDs ;* ***28b)-*** *arm containing the photodiode*

Then a flexible cable connects the probe with the breadboard where is assembled the LED Driver, the Photodetection and Timing Circuits.

3.4.1 . Modules

This section contains the details of the circuits used to assemble the oximeter probe, as well as the problems encountered and their solutions. The circuits were assembled on a breadboard (Figure 30) to be easier to choose the values of electronic components. The oximeter probe is constituted by three main modules: LED driver, photodetection and timing modules.

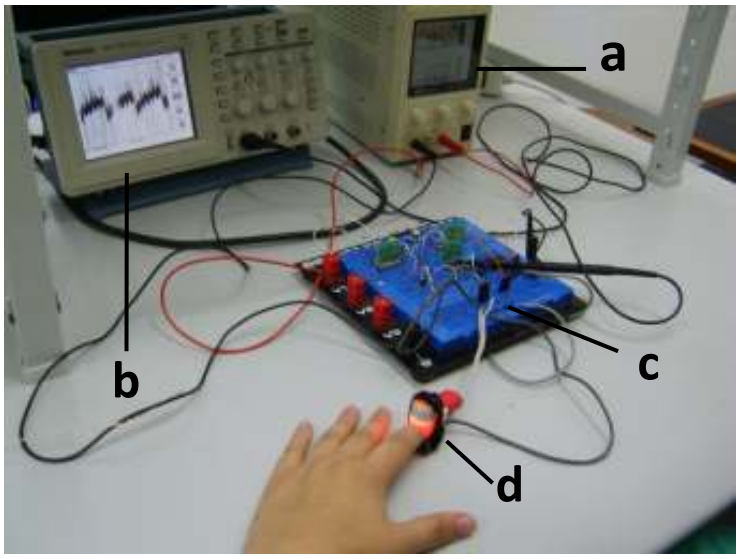


***Figure 29**. Workbench at GEI; a – power supply; b – oscilloscope; c – breadboard containing the circuits ; d – oximeter probe*

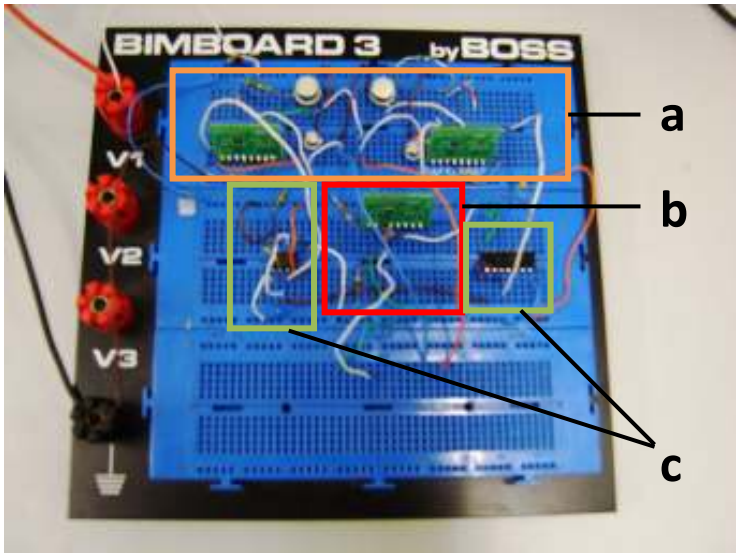


***Figure 30**. Breadboard containing the modules; a – LED driver circuit; b – photodetection circuit; c – timing circuits*

3.4.1.1. LED Driver Module

The circuit was implemented through the use of LEDs that operate in the range specified in the absorption curve (Figure 4) and to benefit from the maximum differences between the spectra of Hb and HbO_2. It was used red LEDs of 635nm [36] and infrared LEDs of 950nm [37].

To prevent low levels of signal detection through the finger, it was chosen ultrabright LEDs, which have an internal system of lenses that provide high-brightness output. In order to ease the detection process and since there are light sources that are multiplex in time in discrete event and only one photodetector, LEDs were chosen for high-current peak, in order to increase peak power [21].

The optical power of the LED increases in an approximately linear way with intensity of current. Thus, the polarization circuit of the LEDs should ensure that during the operation of each of the LEDs, the current which passes through them will be constant to ensure a constant optical power. Furthermore, the circuit should be projected so that the red and infrared LEDs operate alternately [10].

Based on the work developed on GEI which was described on Section 1.3., it was developed first the polarization circuit of red LED. The circuit is represented in Figure 31 [10].

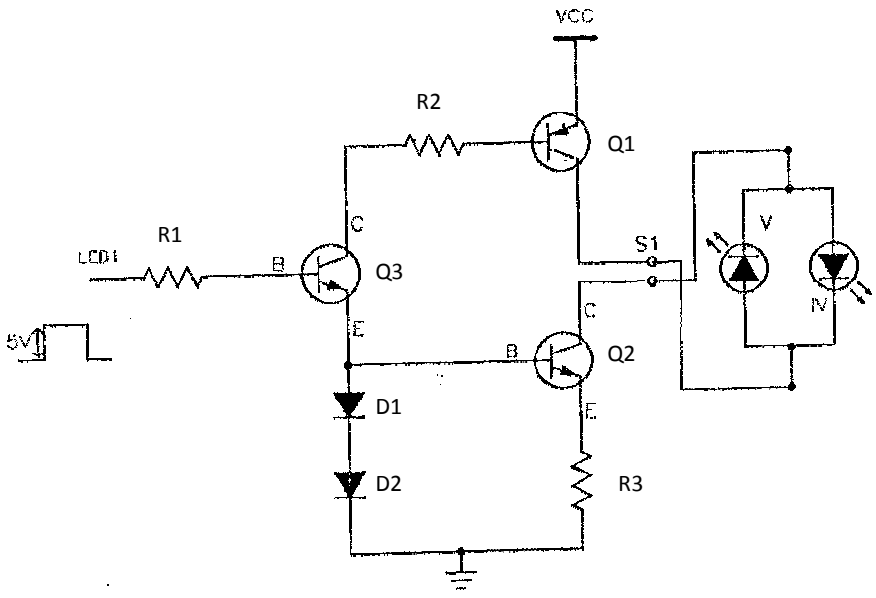


Figure 31.*Polarization circuit of the red LED.* Adapted from [10].

This circuit includes (besides the LED) three transistors Q1 [22], Q2 [23] and Q3 [24] that are used to control the current in the circuit; two diodes D1 and D2 [25] which control the voltage in the base of transistor Q2, that will be 1.4V; and the resistors R1, R2 and R3. It was adopted the same values for resistors R1, R2 and R3 as those used in the work [10]. So R1=5.6 kΩ, R2=560 Ω and R3 =10 Ω.

Using this configuration for the polarization circuit of the red LED, it was possible to view at oscilloscope good biological signals and according to what is expected.

The switching between red and infrared channels will be done through a square wave between 0 and 5V. On the next paragraphs it will be described what happens when it the input of the circuit is applied a 0 or 5V voltage

When a voltage of 5v is applied in the input of the circuit, Q3 is on the linear region with base voltage V_{b3} of 2.1V, once the voltage in the base of transistor Q2 is 1.4V and the base – emitter voltage V_{BE} of a transistor in its active linear region is 0.7V. The transistor Q2 is also in the linear region and, as the resistor R3 is 10Ω, it has an emitter current I_{e2} around 70mA. Since the emitter current of a transistor is approximately equal to the collector current I_c, then the current through the red LED is approximately 70mA.

In the case of transistor Q1, which is a PNP transistor, where the collector–base voltage V_{CB} is greater than zero and the emitter-base voltage V_{EB} is also greater than zero, Q1 is saturated. As the collector-emitter voltage V_{CE} is 0.4V [22], the emitter voltage of Q1 V_{e1} is 4.6V.

The LED, when is driving, has a voltage drop around 2V. As LED voltage is equal to the collector voltage of Q2 V_{c2} and this one is on the linear region, so $1.4\ V < V_{c2} < 2.6\ V$.

In the case of that is applied a voltage of 0V in the input of the circuit, the base-emitter voltage is less than zero and the transistor Q3 cut-off. Then I_{c3} and I_{b3} are zero and Q1 and Q2 cut-off too and there is no current through the red LED.

As it is possible to conclude by the previous analysis, the resistor R3 is the responsible by the value of current which passes through the LED, so it affects the intensity of light. So in the chosen of component values of the polarization circuit of the infrared LED, only resistor R3 was changed; the other components remained with the same values.

Due the super bright presented by the infrared LED, it became clear that the current through it should be lower so that not to saturate the photodiode. Thus, the value

of the resistor has been gradually increased until it is possible to see good signals at the oscilloscope. It occurs for a resistance of 50 Ω.

Repeating the analysis done to the red LED, when in the input of the infrared LED polarization circuit is applied a voltage of 5V, as the voltage in the base of transistor Q2 still is 1.4V, the current through the infrared LED is approximately 14mA. When the input of the circuit is applied a voltage of 0V, the situation is the same of red LED circuit and there is no current through the LED.

Both circuits described above which constitute the LED Driver Module were tested on the tool Multisim and are represented on the Attachment E. With this circuit it will be possible to switch the LEDs, through a timing module developed on the Section 3.4.1.3: when in the input of the polarization circuit of the red LED is applied a voltage of 5V, the red LED is on; when the voltage passes from 5 to 0V, the LED turn off and in the input of the polarization circuit of the infrared LED the voltage passes from 0 to 5V and the infrared LED is on.

By the analysis of the circuit on the Attachment F and through experimental measurements, it is possible to conclude that:

- the current which passes through red LED is around 70mA (when the LED is on)
- the current which passes through infrared LED is around 14mA (when the LED is on)
- at the moment that the polarization circuit of each LED stops driving the respective LED, there is no current passing to the other one

3.4.1.2. Photodetection Module

The main components of the photodetection module are the photodiode and the op-amp used as current-voltage converter. According to the literature and using part of the circuit of the Attachment C (that was used on the development of the first oximeter probe of the project), Figure 32 represents the transimpedance amplifier configuration with a photodiode of the photodetection module:

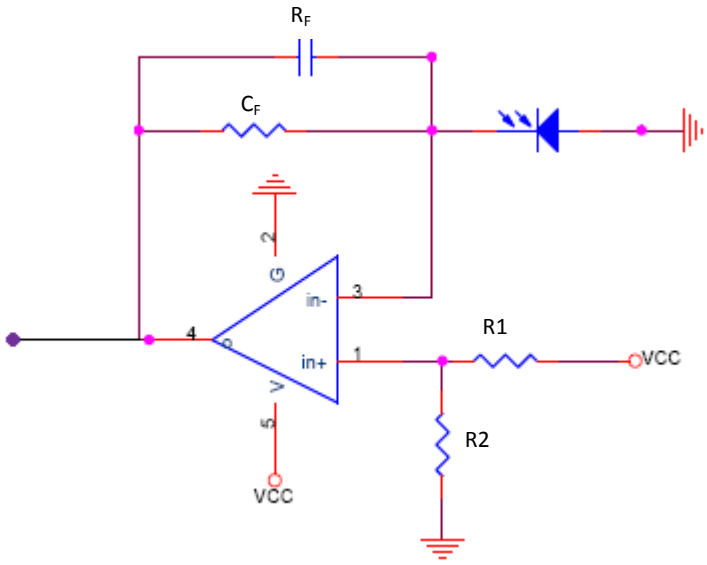


Figure 32. *Transipedance amplifier configuration with a photodiode*

The configuration of Figure 32 has a difference from the presented on the Figure 10 because the input 1 is not connected to the ground. The resistors R1 and R2 have the same value to ensure that the voltage in V+ is 2,5V and the signal is around this value. If the input was grounded, it would be possible to see only the positive part of the signal detected since the op-amp is powered to 0-5V. Thus, it is "up" the level of reference from 0V (ground) to 2.5V and so the signal is seen around this value. So R1 and R2 have the same value which is 10KΩ.

The photodiode used should have satisfactory answers to both wavelength (red and infrared), since it will use only one photodiode for both wavelengths, so it should has a sensitivity spectral range that includes the wavelengths used [21].

Moreover, in the choice of the photodiode some considerations should be taken into account [1]:

- **Photodiode Capacitance** – the junction capacitance affects noise and the bandwidth of the circuit so it should be as low as possible;
- **Photodiode Active Area** – the area of the photodiode is directly proportional to the junction capacitance so the photodiode active area should be as small as possible for largest signal-to-noise ratio.

Due the considerations done above, the photodiode used on the project is the SILONEX - SLCD-61N1, whose the main specifications are exposed on the Table 1 below:

Sensivity	0.55A/W	
Active Area	10.40 mm²	
Junction Capacitance	0.4 nF	
Dark Current	1.7µA	
Sensitivity Spectral Range	400 nm – 1100 nm	

Table 1. *Main features of the photodiode SILONEX - SLCD-61N1.* From [26]

The photodiode generates a current proportional to the intensity of light and then the referred transimpedance amplifier was used to convert current into voltage. These are the most common types of amplifiers used in pulse oximetry application nowadays. For the amplification of signal, the current of the photodiode should be converted into voltage with moderate impedance [21].

The op-amp used on this module was the NATIONAL SEMICONDUCTOR - LM321MF, with the main specifications are exposed on the Table 2 below:

Gain, Bandwidth -3dB	1 MHz	
Input Capacitance	100 pF	
Voltage, Supply Min	3V	
Voltage, Supply Max	32V	
Slew Rate	0.4	

Table 2. *Main features of the NATIONAL SEMICONDUCTOR - LM321MF.* From [27]

The transimpedance amplifier configuration is the most frequently used in pulse oximetry applications but it has a number of multidimensional constraints so some important considerations must be taken in account on the chosen of the components [1]:

- **Feedback Resistor** – the feedback resistor should be made as large as possible to minimize the noise because it is the main source of the noise in the circuit
- **Feedback Capacitor** – this component improves stability and minimizes gain peaking; a general formula for the choice of the capacitor value is [1]:

$$C_F = \frac{1}{4\pi R_F f_c}\left(1+\sqrt{1+8\pi R_F C_I f_c}\right) \qquad \textbf{(Equation 20)}$$

where

- f_c is the unity gain frequency of the op-amp;
- C_I is the total input capacitance (photodiode junction capacitance + op-amp input capacitance);
- R_F is the feedback resistance.

As was previously mentioned the feedback resistor should be as large as possible. Through some experiences and consulting some references [1], [13], it was chosen the value 330K for the feedback resistor as a suitable value. Using the Equation 20 to determine the value of feedback capacitor:

$$C_F = \frac{1}{4\pi R_F f_c}\left(1+\sqrt{1+8\pi R_F C_I f_c}\right)$$

where:

$R_F = 330\text{x}10^{3}$ Ohm

$f_C = 1\text{x}10^{6}$ Hz

$C_I = 0.4\text{x}10^{-9} + 100\text{x}10^{-12} = 5\text{x}10^{-10}$ F

$$\mathbf{C_F \approx 15\ pF}$$

The feedback capacitor used has the value of 15 pF, because it was the closest existing value .

Now it is possible to calculate the bandwidth by:

$$BW = 1{,}4\ f_P \qquad \textbf{(Equation 21)}$$

where

$$f_P = \sqrt{\frac{f_c}{2\pi R_F (C_I + C_F)}} \qquad \textbf{(Equation 22)}$$

so

$$f_P \approx 30{,}6 \text{ kHz}$$

and

$$BW \approx 43 \text{ kHz}$$

Since it was expected frequencies around few Hz, the bandwidth of the circuit is adequate to receive the biological signals. The inconvenient is that the bandwidth can be too large and so the signal may be affected by noise. However this is a drawback that it is considered acceptable and suitable for software processing.

3.4.1.3. Timing Module

The BioPlux™ module has only one digital output port so it is not possible to multiplex the LEDs through it. Therefore, it was necessary to develop an electronic circuit on the breadboard that could be connected to the driver circuit of the LEDs and proceed to the timing of them. The timing circuit was built around the 555 timer and the NAND gate. The 555 timer is used to produce a TTL (Transistor-Transistor Logic) signal, while the NAND gate is used to invert the TTL signal through a NOT.

At the Attachment G is possible to find the complete LED Driver circuit including the timing modules for each LED.

The 8-pin 555 timer is an integrated circuit (IC) and can be implement in a variety of timer and multivibrator applications, which becomes useful in many projects [29].

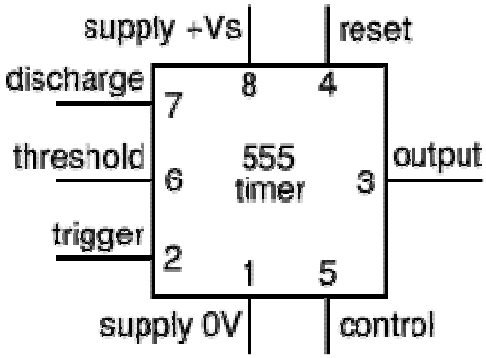


Figure 33. *An 8-pin 555 timer.* From [28].

The 555 has three operating modes: bistable mode (a simple memory which can be set and reset); monostable mode (producing a single pulse when triggered, so it functions as a "one-shot") and an astable mode (producing a single pulse when triggered so it works as an oscillator). In this project, it was used the astable mode [28], [29].

An astable circuit outputs rectangular pulses which constitute a digital waveform with sharp transitions between low (0V) and high (+Vs), with a specified frequency, as it is possible to see in Figure 34.

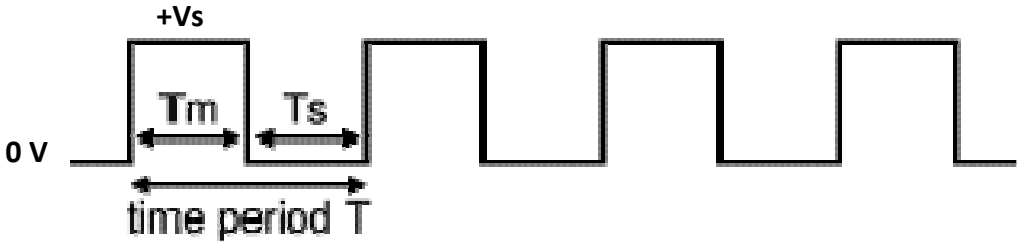


Figure 34. *555 astable output: a square wave.* Adapted from [28].

The time period (T) of the square wave is the time for one complete cycle, which corresponds to a frequency (f) which is the number of cycles per second. The time period can be split into two parts: mark time (Tm) when output is high and space time (Ts) when output is low [28].

In Figure 35, it is represented a standard 555 timer circuit.

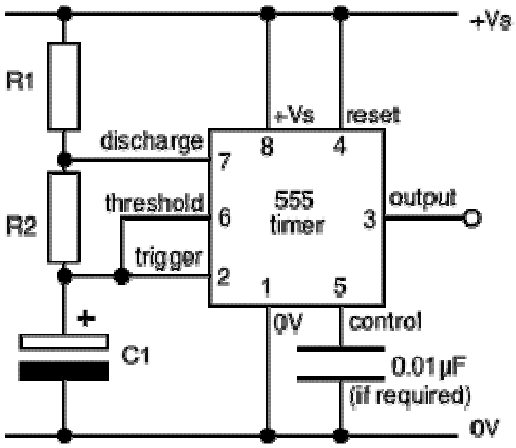


Figure 35. *555 astable circuit.* From [28]

The resistor R1 is connected between Vs and the discharge pin (7) and R2 is connected between the discharge pin (7) and the trigger and threshold pins (2 and 6 respectively) that share a common node.

In this circuit [28]:

$$T = 0{,}7\ (R1 + 2R2)\ C1 \qquad \textbf{(Equation 23)}$$

$$f = \frac{1{,}4}{(R1 + 2R2).C1} \qquad \textbf{(Equation 24)}$$

where:

- T is the time period in seconds (s)
- f is the frequency in Hertz (Hz)
- R1 is the resistance in Ohms (Ω)
- R2 is the resistance in Ohms (Ω)
- C1 is the capacitance in Farads (F)

The 555 astable circuit operation is represented in Figure 36. With the output high, the capacitor C1 is charged by the current flowing through R1 and R2. The threshold and trigger inputs monitor the capacitor voltage and when it reaches 2/3 Vs (threshold voltage) the output becomes low and the discharge pin is connected to 0V. The capacitor

discharges only through R2 into the discharged pin (7). When the voltage falls into 1/3 Vs (trigger voltage) the output becomes high again and the discharge pin (7) is disconnected, allowing the capacitor to start charging again.

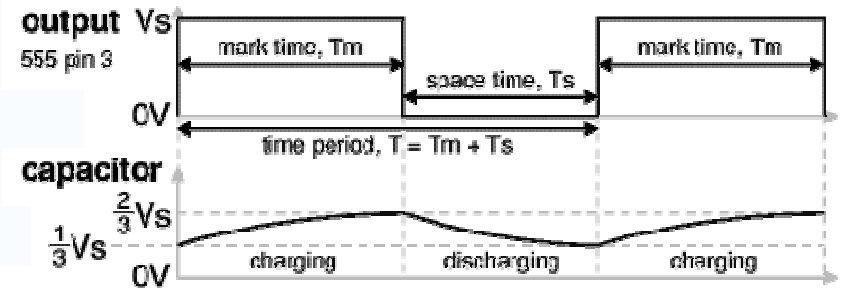


Figure 36. *555 astable circuit operation.* From [28]

As said previously, the time period (T) can be split into two parts: T_m and T_s, which are function of R1 and R2, as shown in Equation 25 and Equation 26 [28]:

$$T_m = 0.7\ (R1+R2)\ C1 \qquad \textbf{(Equation 25)}$$

$$T_s = 0.7 \times R2 \times C1 \qquad \textbf{(Equation 26)}$$

In the development of the timing module and in order to switch the LEDs, it is intended that T_m and T_S are almost equal. So the duty cycle (the percentage of the complete cycle for which the output is high) should be almost 50%, which is achieved if R2 is much larger than R1, as it is possible to verify in Equation 26.

$$\text{Duty Cycle} = \frac{T_m}{T_m + T_s} = \frac{R1 + R2}{R1 + 2R2} \qquad \textbf{(Equation 26)}$$

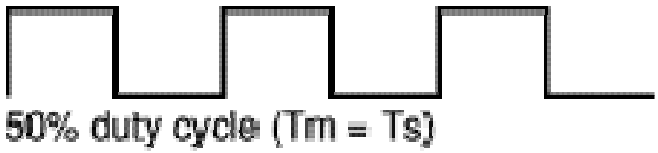


Figure 37. *A 50% duty cycle square wave.* From [28]

In order to choose the values of circuit components, it is necessary to choose R1, R2 and C1. In Attachment H it is available a 555 astable frequencies table which can be used as a guide in the choice of the components. The maximum cardiac frequency is never

going to be more than a few Hz. The sampling frequency should be well above the maximum cardiac frequency present in the arterial pulse to get a better defined curve where can be detected the events and because the curve of the arterial pulse has much more components (besides the heartbeat) with higher frequencies. So, taken into account the expected frequencies of the biological signal and after experimental tests, it was concluded that a frequency around 680Hz is suitable so it was chosen C1=1nF and R2=1MΩ. Using Equation 24 and admitting that R2 is much larger than R1 (because it is intended $T_m=T_s$), the frequency f can be written as:

$$f=\frac{0.7}{R2\,x\,C1} \qquad \textbf{(Equation 27)}$$

so,

$$f = 700\ Hz$$

R1 should be about a tenth of R2 in order to be ignored. So it was chosen the value of 100 kΩ.

Now, using Equation 26, it could be calculated the exact value of duty cycle:

$$\text{Duty Cycle} = 52{,}4\% \approx 50\%$$

In the next table, it is presented the specifications apply to the 555 timer used in the project.

Supply voltage (V_{CC})	5 to 15 V
Supply current (V_{CC} = +5 V)	4.5 to 16 V
Output current (maximum)	200 mA
Operating Temperature	0 to 70 °C
Power dissipation	600 mW

Table 3. *Main features of the NE555-timer.* From [30]

The astable circuit was connected with the red LED driver circuit. In order to allow that the infrared LED flash off when red LED flash on and vice versa, the output of the astable circuit was connected to a NAND gate.

A NAND gate is a digital logic gate and is a combination of the digital logic AND gate with an inverter or NOT gate. These are connected together in series, as shown in Figures 38 and 39.

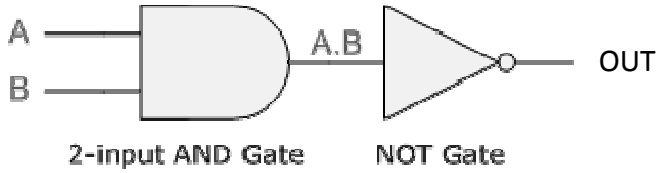


Figure 38. *Digital logic AND gate with that of a NOT gate connected together in series.* Adapted from [31].

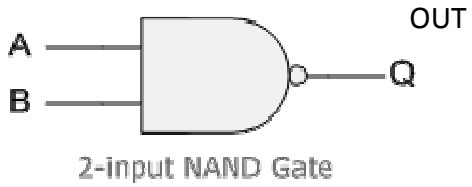


Figure 39. *2-Input NAND Gate.* Adapted from [31].

The NAND gate will behave in a manner that corresponds to the truth table 4.

INPUT		**OUTPUT**
A	B	A NAND B
0	0	1
0	1	1
1	0	1
1	1	0

Table 4. *Truth Table of a NAND gate.* Based on [31].

As it is possible to see in Table 4, a low output results only if both the inputs to the gate are high; if one or both inputs are low, a high output results. So, if the inputs to the gate are shunted, the NAND gate will return only two results. If the input from 555 timer circuit is low, both inputs to the gate will be low and it results in a high output. In the opposite, if the input from 555 timer circuit is high, both inputs to the gate will be high and it results in a low output.

The NAND gate is powered by a VCC of 5V which will be corresponding to the high output returned and is connected also with ground which corresponds to the low output returned.

In Figure 40, it is possible to see at the oscilloscope the signal at the output of the 555 timer and the NAND gate:

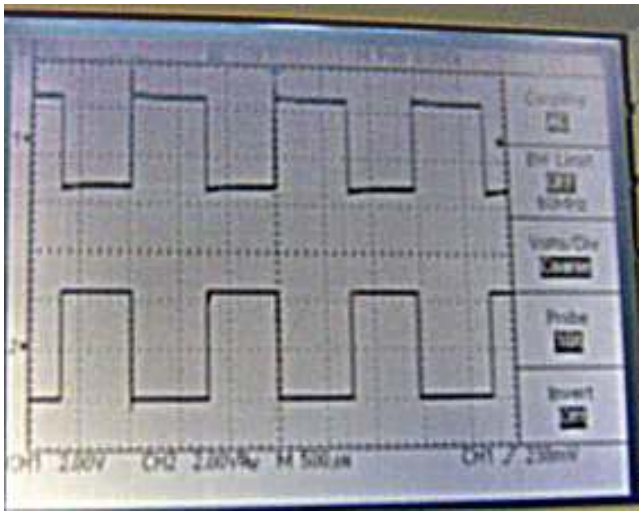


Figure 40. *Signal at the output of thet 555 timer and the NAND gate seen at the oscilloscope*

The first curve of Figure 40 corresponds to the output of 555 timer and the second curve corresponds to the output of NAND gate. In Figure, it is clear that both signals are lagged in time.

The output of the NAND gate is connected with the infrared LED driver circuit and now the both LEDs are multiplex.

3.5. Data Acquisition Platform

A Data Acquisition (DAQ) device is responsible for collecting information such as signals in order to generate data that can be manipulated for instance by a computer.

The NI-6009 is DAQ device with 8 analog inputs (14-bit), 2 analog outputs (12-bit) 12 digital I/O and 32-bit counter. With plug-and-play USB connectivity, these modules are simple enough for quick measurements [33].

Figure 41. *NI-6009 DAQ*

The manufacturer provides an interactive software of control called LabVIEW SignalExpress which can acquire, monitor, analyze and store the data.

In this project, NI-6009 board was connected with the output of the photodetection module and through its software, it is possible to acquire the signals with an acquisition frequency, view it in real time and record the data; with the possibility of viewing the biological signals provided by the oximeter probe, it helps to evaluate the quality of the signals acquired. The data is than stored as a text file in order to be processed by the algorithms developed in the processing module. The frequency of acquisition will depend on the sampling and the kind of events that are supposed to view.

Thus, at that time NI-6009 DAQ is only an intermediate platform of work to enable the acquisition of data for the development of the oximeter probe and for the evolution of the detection algorithm to implement in the final platform of the product in the future.

3.6. Data Processing Tools (Matlab)

Matlab (Matrix Laboratory) is an interactive software geared for the numerical calculation. Matlab integrates numerical analysis, calculus with matrices, signal processing and construction of graphs in an easy way to use.

As it was said in Section 3.5., the data is stored as a text file by the NI-6009 DAQ. In order to evaluate the quality of the signal, it was developed a program in Matlab which allows viewing the signals acquired later.

The program developed is able to make a simple processing of the signal through MATLAB's own functions in order to reduce noise and thus enable the identification of the peaks of the signal to calculate the heart rate. This algorithm does not interfere with the process of the signal by the acquisition and process module; it is used only for evaluate the quality of the signal and identify the heart beat pattern.

The algorithm implemented in Matlab can be found in the Attachment D. The algorithm can be represented by the flowshard of Figure 42 and it will be explained in the next paragraphs.

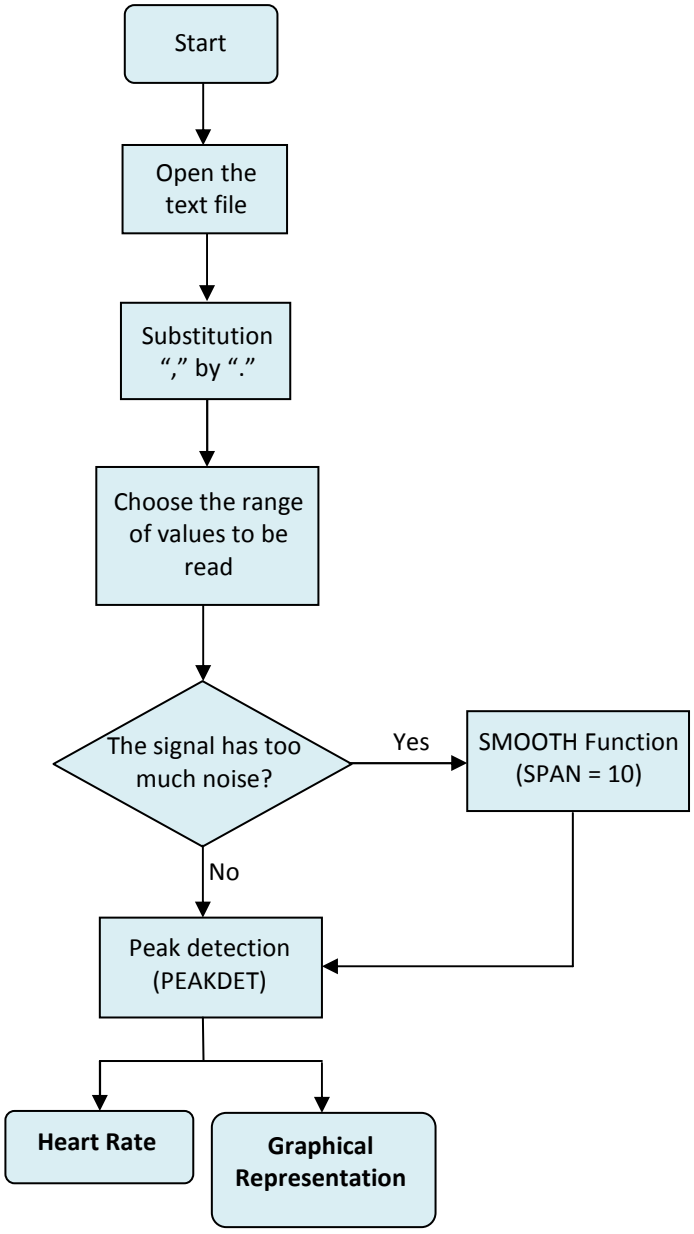


Figure 42. Flowshard of Matlab algorithm

First the program opens the text file and the values are converted into values that can be read by Matlab. Then it is established the range of values that will be printed. The function PEAKDET is used to identify the maximum and minimum of the previous curve. This function detects peaks in a vector, finding the local maxima and minima in it. Then others vectors are created (MAXTAB and MINTAB) where one column contains the X-values and the other the found value corresponding to a peak. A point is considered a maximum peak if it has the maximal value and was preceded (to the left) by a value lower than DELTA. DELTA is a variable which should be at least equal to the maximum difference of amplitude that can be mistaken as peak. This function was developed by Eli Billauer [34]

and it is possible to find it in the Attachment E. In order to avoid the detection of false peaks, the value of DELTA is chosen according to the signal acquired and through the visual analysis of the graphical representation of it.

Then it is calculated the length of the vector returned by the function PEAKDET using 10s of the signal detected and the result is multiplied by 6 in order to estimate the number of maximum in one minute that which corresponding to the heart rate.

However, many times the original signal presents a lot of noise, so it is done a simple processing of the signal through the Smooth (Y, SPAN) function of Matlab, which is based in the moving average method. This function works as a filter since it smooths data Y using SPAN as the number of points used to compute each element of the new data. In the project, where it was necessary to apply the Smooth function, it was used a span of 10, which means that the moving average method is applied with span 10.

As in the case of the original signal, the algorithm to detect peaks can be applied to this new signal and thus it is possible to calculate the heart rate in a similar way to that described above.

Finally, the graphs are represented and the peaks are detected with the maximums at red and the minimums at green.

4. RESULTS/ DISCUSSION OF RESULTS

4.1. Tests

Three tests were done to evaluate the performance of the probe developed and the implemented algorithms. The tests were performed on a 23 years old female patient, weighing 63kg and 1.70m tall. The patient does not practice sport regularly and has no history of cardiovascular or respiratory diseases.

The signals were acquired by the NI-DAQ and stored as a file text. The frequency of acquisition used depends of the test that was done. In the case of a test where the LEDs are used in a continuous way and taking in account that the frequency present in the arterial blood is around few Hz, it is used the frequency of acquisition of 800Hz. This value is well above the maximum frequency present and experimental tests proved that this frequency of acquisition is suitable to detect the events

In a case that the LEDs are multiplexed with a sampling frequency of 700Hz (Section 3.4.1.3.), the frequency of acquisition could be chosen according the Sampling Theorem [6], [35]:

$$f_{acq} \geq 2\, f_s \qquad \textbf{(Equation 28)}$$

where,

- f_{acq} is the frequency of acquisition of the signals
- f_s is the sampling frequency

This condition is necessary to ensure that the expected events will be possible to see and no important information will be lost. So, as the sampling is 700Hz, in the project it was used a frequency of acquisition of 1.4 kHz.

The first test was performed in a situation where the patient was at rest before the measurements. In order to assess the signal quality and performance of the oximeter, it was done some variants on the tests:

- 1A - inserting the finger in the oximeter probe and using only the red LED in a continuous way;

- 1B - inserting the finger in the oximeter probe and using only the infrared LED in a continuous way;
- 1E - inserting the finger in the oximeter probe and multiplexing the both LEDs.

The second test was performed through the simulation of physical activity by the patient, through the ascending and descending stairs for 3 minutes before the measurements. On this test, it was done some variants too:

- 2A - inserting the finger in the oximeter probe and using only the red LED in a continuous way;
- 2B - inserting the finger in the oximeter probe and using only the infrared LED in a continuous way;
- 2C - inserting the finger in the oximeter probe and multiplexing the both LEDs.

It was also intended to perform a third test that simulates a state of apnea by holding breath. However for this test to be valid and for it really simulate a situation of apnea it must be done in a medical environment with the amount of inhaled oxygen being monitored. So the test of the pulse oximeter performance in that situation should be done in the future.

The obtained results and its analysis can be found in the next section.

4.2. Final Results

Once the signals were acquired and stored, the algorithms developed to be implemented in the processing module were applied to it, in order to provide the heart rate and the ratio between minimums and maximums. Moreover, to evaluate the quality of the signals, they were printed in Matlab and an estimative of the heart rate was made through a simple program. On the detection of the peaks, DELTA of Matlab's function PEAKDET was chosen to be at least equal to the maximum difference of amplitude that can be mistaken as peak, such as peaks due to the noise or false peaks resulting from physiological phenomenon. The value was validated from visual analysis of the graphical representation of the acquired signals. In this section, it will be presented the results obtained with the tests done to evaluate the performance of the probe and their discussion.

To better understand the shape of the curves that will be shown, it is important the analysis of Figure 43 which is represented a portion of a graphical representation of a signal.

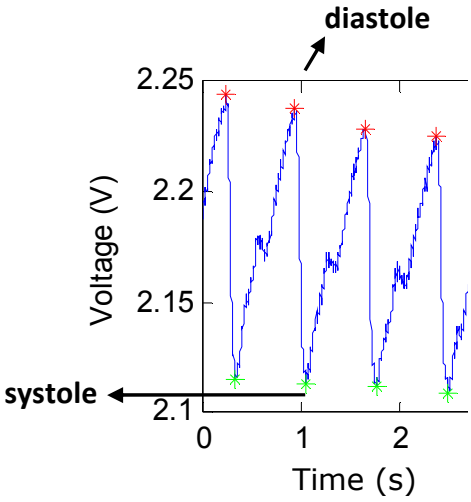


Figure 43. *Portion of a signal graphical representation*

In systole, the arterial blood volume is the greatest, so the peripheral blood volume in the capillaries is also the greatest. For finger transmission oximeter, this means that at this time there will be a light transmission minimum (Figure 43). On the other hand, the arterial blood volume is the lowest in diastole so the peripheral blood volume in the capillaries is also the lowest at that time which can be seen as a light transmission

maximum in the graphical representation of the signal (Figure 43). The sawtooth notch that is possible to see in Figure 43 can be assumed as the dicrotic notch that occurs between systole and diastole.

- **1A**

In Figure 44 it is possible to see the first ten seconds of the signal acquired using a red LED in a continuous way.

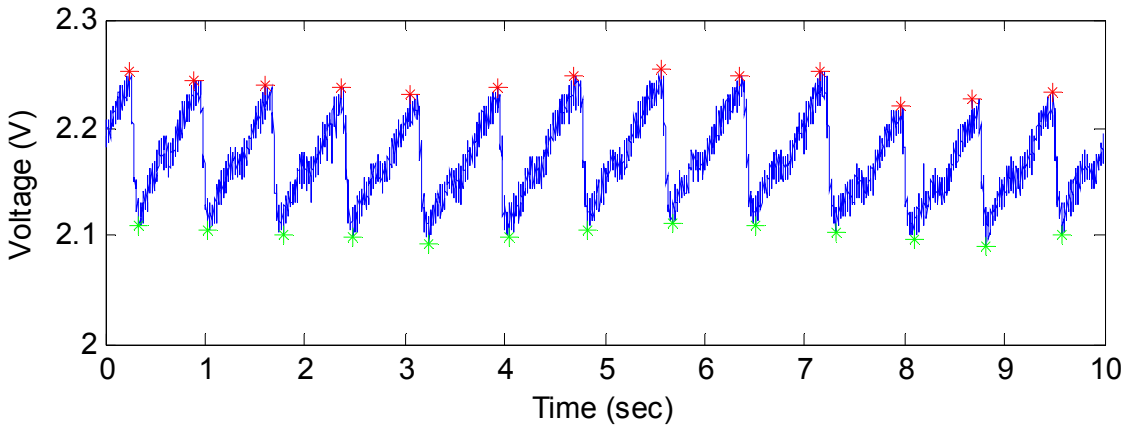


Figure 44. *graphical representation of the original signal acquired with NI-6009 DAQ, between the 0 and 10 sec using for the test 1A*

The signal of Figure 44 has an acceptable noise and it is clearly possible to see a pulse signal. It is getting voltages between 2.1V and 2.25V (signal is around 150 mV).

It is important to note that the trace in Figure 44 has a structure that was assumed as the dicrotic notch. As said in Section 2.3., it can cause signal processing problems with the erroneous detection of another peaks. So, analyzing the graphical representation, the occurrence of this phenomenon is the main criterion to the choice of the value of DELTA which is 0.06V.

In Figure 44 is possible to see the detection of the maximums at red and the minimums at green. Based on it, the algorithm developed in Matlab returns a value of heart rate 78bmp, for this test.

In order to get a better signal and the algorithm of peak detection works better, it was applied to the signal of Figure 44, the program developed in Matlab based on function Smooth (Span = 10) . After the treatment by Matlab it is obtained the curve of Figure 45.

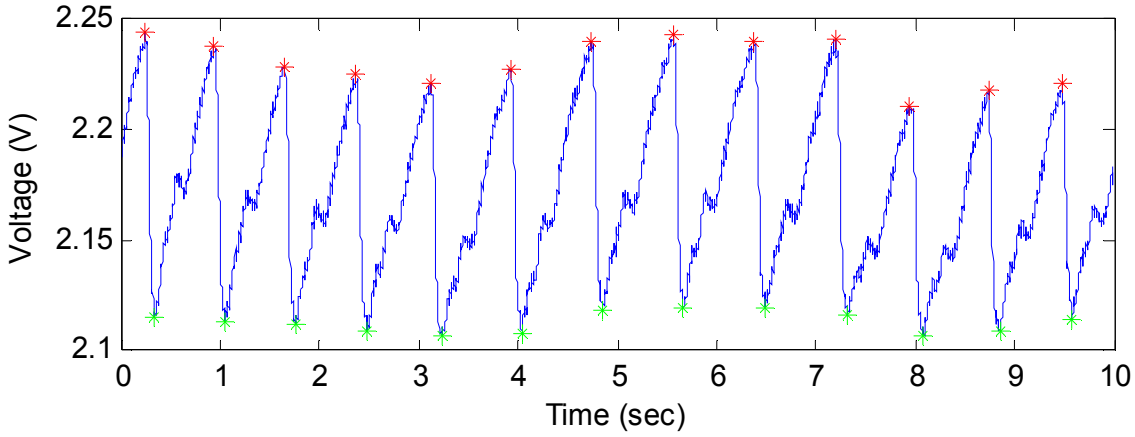


Figure 45. *graphical representation of the signal acquired with NI-6009 DAQ, between the 0 and 10 sec to the test 1A, after Matlab processing*

In Figure 45 the signal is clearly better defined and is visible the pulsatile pattern and the dicrotic notch. Due to the moving average, the curve of the figure is flatter but this fact does not interfere with the estimation of the heart rate but with the choice of the value of DELTA, which needs to be less. Due to the dicrotic notch, the value of DELTA used is 0.015V. Now, it is done the detection of the maximums and minimums and the value of heart rate returned is 78bpm, which coincides with the signal of Figure 45.

- **1B**

The first ten seconds of the signal acquired using an infrared LED in a continuous way is shown in Figure 46.

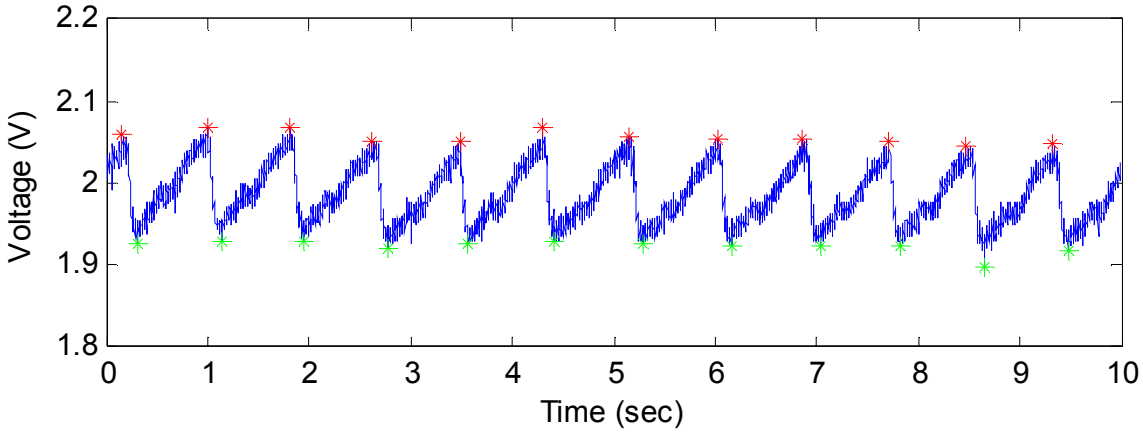


Figure 46. *graphical representation of the original signal acquired with NI-6009 DAQ, between the 0 and 10 sec using for the test 1B*

In Figure 46 it is clearly visible the pulsatile pattern and the dicrotic notch, despite the noise. The voltages are between 1.9V and 2.075V (signal is around 175 mV).

Analyzing the graphical representation, the main criterion to the choice of the value of DELTA is the dicrotic notch again and its value is 0.06V. The maximums are detected at red and the minimums at green and based on it, the Maltlab algorithm returns a value of heart rate 72bmp, for the test 1B.

Figure 47 shows the signal of Figure 46 after the Maltab processing through the function Smooth (span=10).

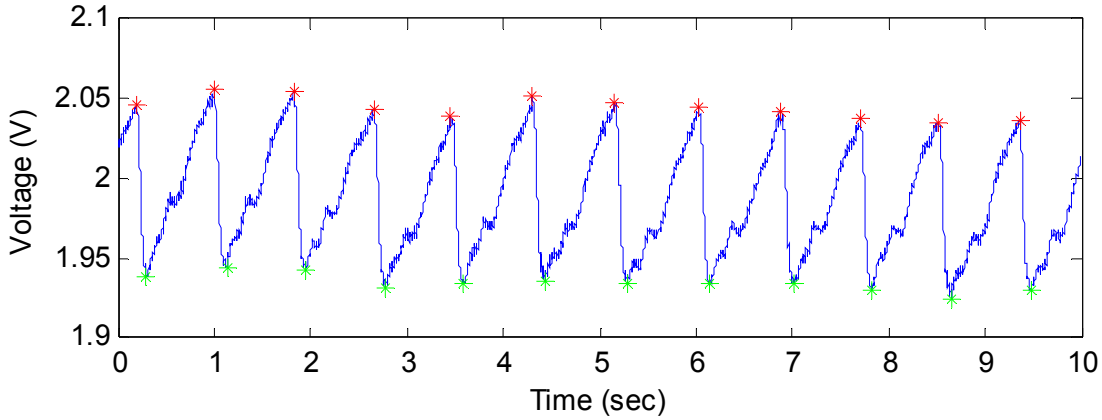


Figure 47. *graphical representation of the signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 1B, after Matlab processing*

As in 1A this curve is better defined but is flatter when compared with the Figure 46. Due to the dicrotic notch which is visible in Figure 47, the value of DELTA used is 0.015V. Through the detection of the maximums and minimums and the value of heart rate returned by Matlab is 72BPM, which coincides with the signal of Figure 46.

- **1C**

In Figure 48 it is possible to see the signal acquired using the both LEDs switched.

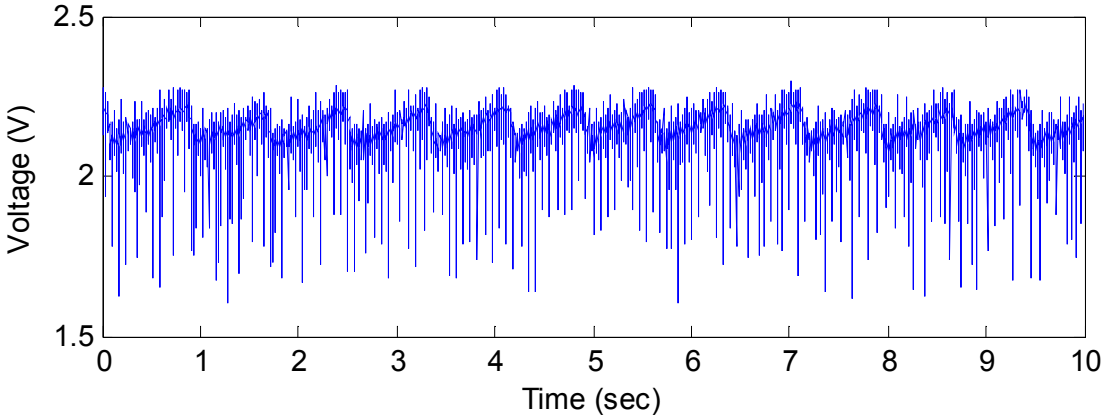


Figure 48. *graphical representation of the original signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 1C*

Due the spectrum of the Figure 48 is a superposition of the red and infrared spectra, the graphical representation presents a lot of interference because the zero value is read many times and the signal return to the activation level of the photodiode. Despite of the interference, it is possible to see a pulse signal which corresponds to the heart beat.

Since this spectrum is a superposition of red and infrared spectra, it makes no sense to apply the function PEAKDET to the signal, since it could occur the detection of false peaks, since each spectrum has its maximum and minimum.

Observing the graph, it seems that this signal has about thirteen peaks in 10 seconds, so the patient has a heart rate around 78bpm.

Table 5 shows the values of heart rate and R_{os} returned by the algorithm developed by the project partner for the test 1C. These values are returned every 10 seconds.

	Heart Rate (bpm)	R_{OS}
0-10 sec	72	1,02627
10-20 sec	78	0,990736
20-30 sec	72	0,994225
30-40 sec	78	1,005511

Table 5. *results returned by the processing algorithm for the test 1C. [39]*

The algorithm separated the spectrum and, after the signal processing, the peaks were detected. Then it was possible to determine the heart rate and the Ratio of Ratios according to the Equation 16 [39}.

- **2A**

In Figure 49 it is possible to see the first ten seconds of the signal acquired using a red LED in a continuous way after a situation of physical effort.

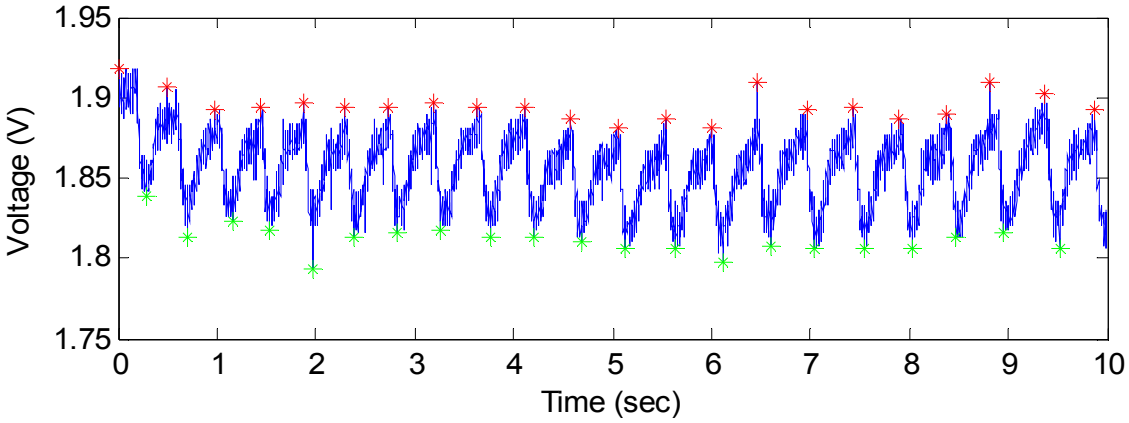


Figure 49. *graphical representation of the original signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 2A*

As it can be seen in Figure 49, the signal of test 2A presents some noise but it is possible to identify a pulse pattern as in the previous situations. It is getting voltages between 1.8 and 1.92V, which means a signal around 182mA.

It was chosen the value of 0.06 V to DELTA and then the peaks are detected (in Figure the maximums are represented at red and the minimums at green) and based on it, the Maltlab algorithm returned a value of heart rate 120bmp, for the test.

To see a better defined curve, the signal is processing by a simple program in Matlab using the function Smooth (span = 10) and the Figure 50 shows the obtained signal.

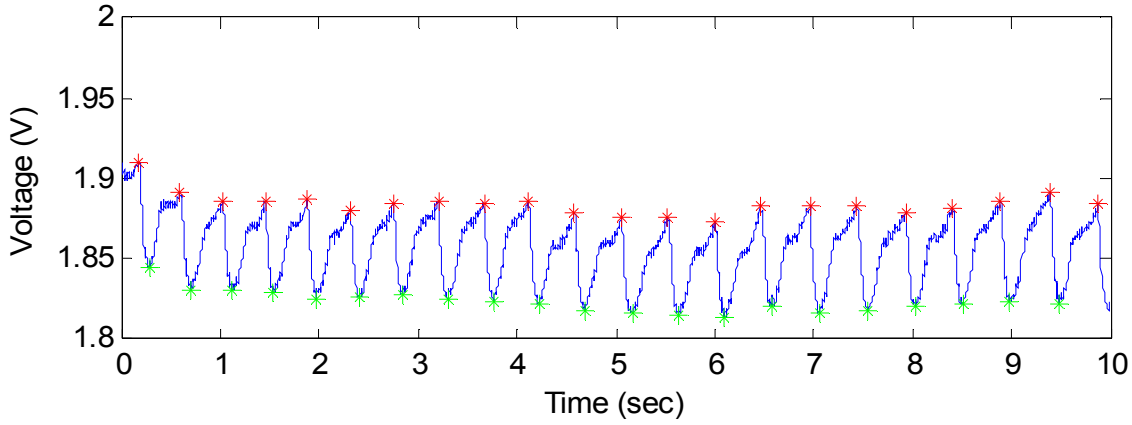


Figure 50. *graphical representation of the signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 2A, after Matlab processing*

In Figure 50 is possible to see that although the curve is flatter than the curve of Figure 49 the dicrotic notch is still visible. The peak detection is applied to the signal and then the program returns the value of heart rate which is 120bpm. This value is consistent with the heart rate to the signal of Figure 49.

- **2B**

Figures 51 shows the first ten seconds of the signal acquired using an infrared LED in a continuous way after a situation of physical effort.

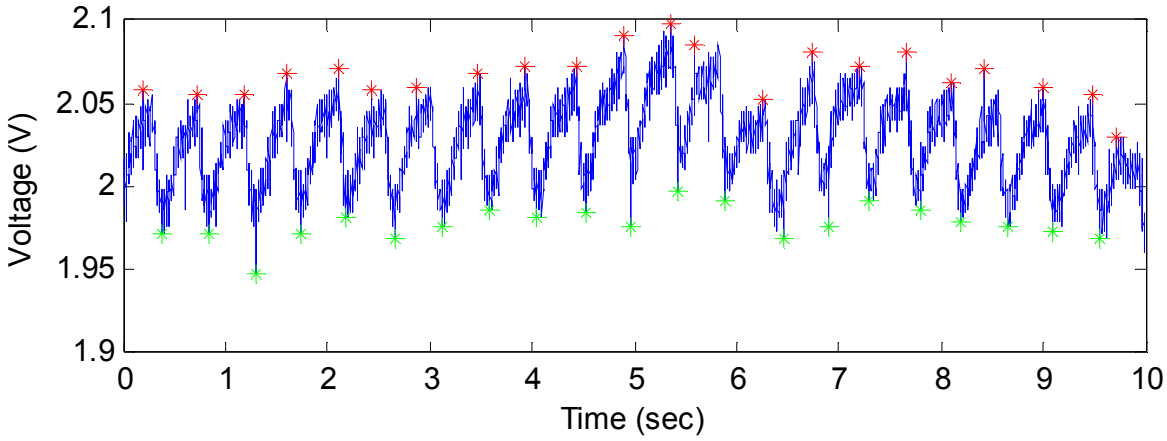


Figure 51. *graphical representation of the signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 2B*

As in previous situations, it is possible to see in Figure 51, a pulse signal corresponding to the heartbeat but it presents a significant band noise. Due to the higher number of beats, is not so clear dicrotic notch that was visible in tests 1A and 1B.

The voltages are between 1.95V and 2.1V which corresponds to a signal around 150 mV. Following the criterion used in test 2A in this test, the value of DELTA is 0.06V. The algorithms of peak detection were applied and Maltlab returns a value of heart rate 132bpm, for the test 2B.

In Figure 52 is presented the obtained signal after the treatment by Matlab using the Smooth function (span=10).

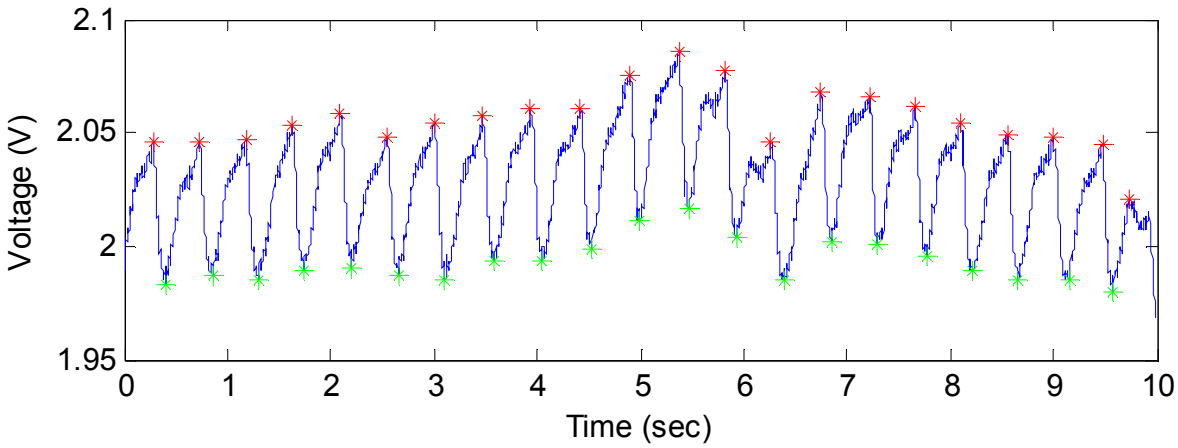


Figure 52. *graphical representation of the signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 2B, after Matlab processing*

In Figure 52, the curve is clearly better defined and is visible the pulsatile pattern. The value of DELTA used is 0.015V. Through the detection of the maximums and minimums, the value of heart rate returned by Matlab is 132BPM, which coincides with the heart rate of the signal of Figure 51.

- **2C**

Figure 53 shows the signal obtained lighting the two LEDs alternately. The measurements were done after a situation on physical effort.

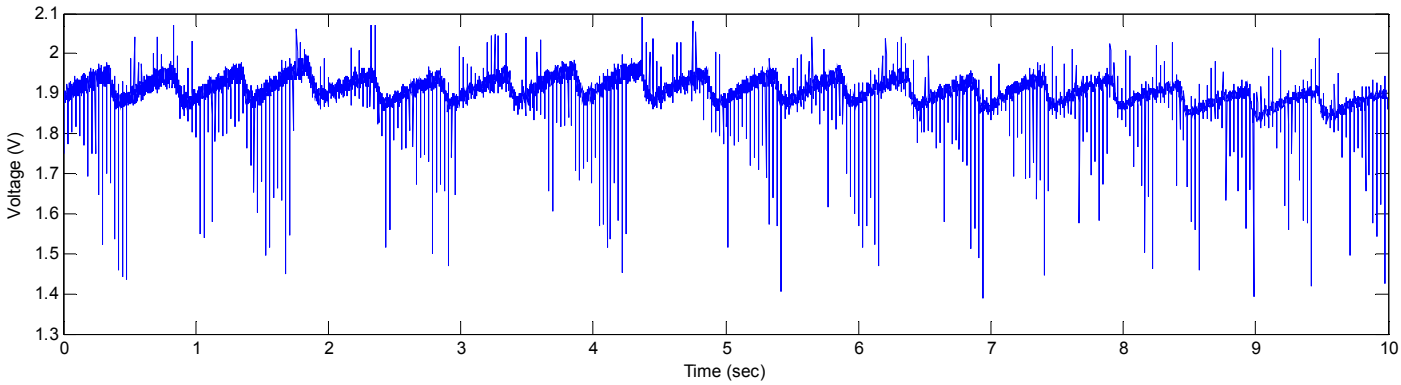


Figure 53. *graphical representation of the original signal acquired with NI-6009 DAQ, between the 0 and 10 sec for the test 2C*

The analysis of the graphical representation of the Figure 53 is similar to the analysis done to the situation 1C. Due the spectrum is a superposition of the red and infrared spectra, the graphical representation presents a lot of interference. Despite of the significant interference, it is possible to identify a pulsatile pattern

Through the observation of the Figure 53, it is possible to identify 19 peaks in the first 10 sec, so it is possible to estimate that the patient presents a heart rate around 114BPM.

Table 6 shows the results returned by the algorithm developed to be implemented in the processing module. These values are returned every 10 seconds.

	Heart Rate (BPM)	R_{OS}
0-10 sec	114	1,005601
10-20 sec	114	1,011159
20-30 sec	114	0,980245

Table 6. *Results returned by the processing algorithm for the test 2C*

The algorithm separated the spectrum and, after the signal processing, the peaks were detected. Then it was possible to determine the heart rate and the Ratio of Ratios according to the Equation 16. [39]

4.3. Discussion of Results

The results presented allow a comprehensive assessment of the performance and efficiency of the probe developed and a comparison with the results obtained by the algorithm developed in C to implement in the processing module, which will return the values of oxygen saturation in blood and the heart rate.

Looking at the graphical representation of the obtained signals and despite the noise, it is possible to indentify clearly a visible pulse signal corresponding to the heartbeat and the physiological phenomenon of dicrotic notch. So it is possible to conclude that the probe and the frequencies used are suitable for signal acquisition in oximetry without loss of relevant information.

For the test carried done at rest (Test 1), the heart rate calculated using Matlab and which can be detected by observation of the curves is between 72bpm and 78bpm. Thus, it is possible to conclude that these values are within normal parameters for situations of absence of physical activity, since the patient does not practice exercise regularly [2].

In a situation of physical activity such as Test 2, it is identifiable a significant increase in heart rate by observing the graphical representations of signals, which has values between 114 and 13bpm. Therefore, it is possible to conclude that the probe is sensitive to these changes in the heart rate caused by physical exercise.

The project partner developed some algorithms in C to be implemented in the acquisition platform [39]. These algorithms allow the reduction of the noise, the separation of red and infrared spectra and then a detection of the signal peaks. For this comparison will only interest tests 1C and 2C which are those in which the algorithm can be applied since the LEDs are switched. Table 5 and 6 show the values of heart rate and R_{os} returned by the program every 10 seconds in order to compare with the values obtained in Matlab and with what is expected.

Regarding to the heart rate between the 0 and 40sec to the test 1C, the program returned values between 72 and 78bpm (Table 5) what is in agreement with the expected values, since the signal analysis in Matlab allows to estimate a heart rate around 78bpm (Figure 48).

For the test 2C, the program returned values of heart rate around the 114bpm (Table 6), coinciding with that ones that were estimated by the analysis of the graphical representation in Matlab (Figure 53).

Comparing Tables 5 and 6, it is possible to verify that the values of R_{os} are quite similar between Test 1 and Test 2. Since with the physical activity the S_pO_2 decreases [4], it was expected than the values of R_{os} of Table 6 are higher than the values of Table 5. Actually, despite the great changes that occur in oxygen consumption and production of carbon dioxide during exercise the mean values of P_{O_2} and pH of the blood remain constant and near the values at rest since the exercise is aerobic [35], as the case of Test 2, where the effort of up and down stairs for three minutes may not be significant to induce changes in the values of saturation.

Comparing the tests using the LEDs on a continuous way with the tests in which the LEDs were powered alternately, it is possible to verify that the last ones have a much noisier signal. This happens because there is not an intermediate state where both LEDs are turned off between the moments in which each LED is lit. Then, the spectrum for each wavelength may have interference caused by light from the other wavelength, which will affect the separation of the spectra and signal processing.

The probe has also proved to be very sensitive to finger movements, a factor which introduced noise in signals acquired, thus affecting the processing of them. Thus, as the probe is not very robust yet, movements of the finger where the oxygen saturation was measured, affect the acquisition by the introducing of noise.

5. CONCLUSIONS AND FUTURE WORK

The Development of a Stand-Alone Pulse Oximeter has been described in the present thesis as a project carried throughout the past academic year.

The technical design of a system able to acquire and process physiological data accurately was the main goal of the project and the central theme of this thesis. Following, the project status is described and analyzed, preceding the presentation of some ideas and suggestions to develop and improve the system's features. Finally a personal appreciation of the overall project is done.

5.1. Project Status

Initially the main objective of the project was to develop a portable pulse oximeter unit that receives data from a probe and processes it in order to return the values of heart rate and oxygen saturation in blood, thus allowing a continuous monitoring of these parameters. As already mentioned in this thesis, the project gained new guidelines and thus the main objectives become the development of an oximeter probe whose data is acquired by BioPlux^{TM} module, which communicate with a platform that would have implemented algorithms in order to perform the signal processing.

At this moment, it is available an efficient probe prototype constituted by economic components which is sensitive to the changes of light that occurs during the arterial pulse. Different measuring conditions such as a situation of physical activity are also detected by the oximeter probe. The oximeter probe developed is able to acquire biological signals to determine the heart rate and S_p0_2. However, the prototype is not robust yet which introduces some artifacts of movement and ambient light that will affect the results.

The functional modules that allow the LEDs driver and switching and the detection of light are also designed, developed and tested. The results show that these circuits are efficient, but still needed several improvements. The circuits were assembled on a breadboard, using a lot of jumpers which will introduce noise in the system.

The BioPlux^{TM} module is ready to acquired data from the probe but it is necessary to adjust acquisition frequency. However, this module has not been tested. This module

communicates via Bluetooth™ with the processing platform, whose the firmware to board initialization and communications protocols was already developed and operational.

The signal processing algorithms were developed in C. They have the ability to apply a filter to the signal in order to reduce noise and then apply a criterion of detection that allows the separation of the spectra, the detection of peaks and thus determine the values of heart rate and the Ratio of Ratios [39].

5.2. Suggestions for a Future Work

In this section some ideas are suggested to continue the project improvement, which can be used as an initial orientation for future developers.

In a short-term, it is really important to assemble a more robust oximeter probe, using the same mechanism of a spring-loaded clip, which allows a better isolation of light and another external noise and decreases the motion artifacts. In order to reduce its size, the new oximeter probe should be assembled with SMD components and have incorporated the operational module circuits (LED driver, photodetection and timing). The construction of the circuit of functional modules in a printed circuit board (PCB) will help eliminate the harmful effects associated with the jumpers used in breadboard.

As it is possible to see in the results obtained, they are affected by some noise so it can be important to develop an electronic filter, which will become the signal processing easier and reliable.

Keeping the energy consumption of the probe in mind, the timing module may be improved by reducing the duration of the duty cycle of each LED, adding a state where the two LEDs are off which will also improve the processing of the signal because the interference caused by the light from the other LED will be reduced.

The system developed must be calibrated so that the results have clinical validity. Thus, it is necessary to perform several tests and from these develop a valid algorithm calibration.

Returning to the original purpose of the work, it seems interesting to the student to take advantage of the hardware and software developed, and design a portable pulse oximeter unit with more autonomy that this one.

5.3. Final Appreciation

The final appreciation of this document regards both the project status and the developed work throughout the academic year.

Regarding the actual status of the project, the student knows that there are a lot of work to do in the future but despite of it, the student thinks that it was done a good work and a low cost oximeter probe was developed and it presents a good performance even though it needs several improvements.

The personal final appreciation of this work is very positive. Technically speaking the student acquired some primarily experience in the electronics field and particularly in circuit construction. As this is a group project, the student acquired a good work team spirit, distributing tasks, sharing ideas and conciliating different perspectives into a final result.

The overall project development allowed a global and practical use of the knowledge acquired though the academic pathway, in such a way that the student hopes to be now more prepared to use her professional and personal skills in the work market.

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Attachment A

The first eleven pages of ISO 9919:2005, containing the Contents, List of Figures, Foreword, Introduction, Scope and Normative References

INTERNATIONAL STANDARD

ISO 9919

Second edition
2005-03-15

Medical electrical equipment — Particular requirements for the basic safety and essential performance of pulse oximeter equipment for medical use

Appareils électromédicaux — Règles particulières de sécurité et performances essentielles du matériel utilisé pour les oxymètres de pouls à usage médical

Reference number
ISO 9919:2005(E)

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Contents

Page

Tables

Figures

Foreword

ISO (the International Organization for Standardization) is a worldwide federation of national standards bodies (ISO member bodies). The work of preparing International Standards is normally carried out through ISO technical committees. Each member body interested in a subject for which a technical committee has been established has the right to be represented on that committee. International organizations, governmental and non-governmental, in liaison with ISO, also take part in the work. ISO collaborates closely with the International Electrotechnical Commission (IEC) on all matters of electrotechnical standardization.

International Standards are drafted in accordance with the rules given in the ISO/IEC Directives, Part 2.

The main task of technical committees is to prepare International Standards. Draft International Standards adopted by the technical committees are circulated to the member bodies for voting. Publication as an International Standard requires approval by at least 75 % of the member bodies casting a vote.

Attention is drawn to the possibility that some of the elements of this document may be the subject of patent rights. ISO shall not be held responsible for identifying any or all such patent rights.

ISO 9919 (IEC 60601-2-54) was prepared jointly by Technical Committee ISO/TC 121, *Anaesthetic and respiratory equipment*, Subcommittee SC 3, *Lung ventilators and related equipment* and Technical Committee IEC/TC 62, *Electrical equipment in medical practice*, Subcommittee SC D, *Electromedical equipment*. The draft was circulated for voting to the national bodies of both ISO and IEC.

This second edition cancels and replaces the first edition (ISO 9919:1992), which has been technically revised.

Introduction

The approximation of arterial haemoglobin saturation and pulse rate using pulse oximetry is common practice in many areas of medicine. This International Standard covers basic safety and essential performance requirements achievable within the limits of existing technology.

Annex AA contains a rationale for some of the requirements. It is included to provide additional insight into the committee's reasoning that led to a requirement and identifying the hazards that the requirement addresses.

Annex BB is a literature survey relevant to the determination of the maximum safe temperature of the interface between a **pulse oximeter probe** and a **patient's** tissue.

Annex CC discusses both the formulae used to evaluate the **SpO_2 accuracy** of **pulse oximeter equipment** measurements, and the names that are assigned to those formulae.

Annex DD presents guidance on when *in vitro* blood calibration of **pulse oximeter equipment** is needed.

Annex EE presents a guideline for **controlled desaturation study** for the calibration of **pulse oximeter equipment**.

Annex FF is a tutorial introduction to several kinds of testers used in pulse oximetry.

Annex GG describes concepts of **pulse oximeter equipment** response time.

This International Standard is a Particular Standard, based on IEC 60601-1:1988, including Amendments 1 (1991) and 2 (1995), hereafter referred to as the General Standard. The General Standard is the basic standard for the safety of all medical electrical equipment used by or under the supervision of qualified personnel in the general medical and patient environment; it also contains certain requirements for reliable operation to ensure safety.

The General Standard has associated Collateral Standards and Particular Standards. The Collateral Standards include requirements for specific technologies and/or hazards and apply to all applicable equipment, such as medical systems, EMC, radiation protection in diagnostic X-ray equipment, software, etc. The Particular Standards apply to specific equipment types, such as medical electron accelerators, high frequency surgical equipment, hospital beds, etc.

NOTE Definitions of Collateral Standard and Particular Standard can be found in IEC 60601-1:1988, 1.5 and A.2, respectively.

To facilitate the use of this International Standard, the following drafting conventions have been applied.

The changes to the text of IEC 60601-1:1988, the General Standard, as supplemented by the Collateral Standards, are specified by the use of the following words.

— "Replacement" means that the indicated clause or subclause of the General Standard is replaced completely by the text of this Particular Standard.

— "Addition" means that the relevant text of this Particular Standard is a new element (e.g. subclause, list element, note, table, figure) additional to the General Standard.

— "Amendment" means that existing text of the General Standard is partially modified by deletion and/or addition as indicated by the text of this Particular Standard.

To avoid confusion with any amendments to the General Standard itself, a particular numbering has been employed for elements added by this International Standard: clauses, subclauses, tables and figures are numbered starting from 101; additional list items are lettered aa), bb), etc. and additional annexes are lettered AA, BB, etc.

In this International Standard, the following print types are used:

— requirements, compliance with which can be tested, and definitions: roman type;

— notes and examples: smaller roman type;

— description of type of document change, and test specifications*: italic type*;

— terms defined in Clause 2 of the General Standard IEC 60601-1:1988 or in this Particular Standard: **bold type.**

Throughout this Particular Standard, text for which a rationale is provided in Annex AA is indicated by an asterisk (*).

Medical electrical equipment — Particular requirements for the basic safety and essential performance of pulse oximeter equipment for medical use

1 Scope

IEC 60601-1:1988, Clause 1 applies, except as follows.

Amendment (add at the end of 1.1):

This International Standard specifies particular requirements for the basic safety and essential performance of **pulse oximeter equipment** intended for use on humans. This includes any part necessary for **normal use**, e.g. the **pulse oximeter monitor**, **pulse oximeter probe**, **probe cable extender**.

These requirements also apply to **pulse oximeter equipment,** including **pulse oximeter monitors**, **pulse oximeter probes** and **probe cable extenders,** that has been **reprocessed**.

The intended use of **pulse oximeter equipment** includes, but is not limited to, the estimation of arterial oxygen haemoglobin saturation and pulse rate on **patients** in healthcare institutions as well as on **patients** in home care.

* This International Standard is not applicable to **pulse oximeter equipment** intended for use in laboratory research applications nor to oximeters that requires a blood sample from the **patient**.

This International Standard is not applicable to **pulse oximeter equipment** solely intended for foetal use.

This International Standard is not applicable to remote or slave (secondary) devices that display $\mathbf{SpO_2}$ values that are located outside of the **patient environment**.

The requirements of this International Standard which replace or modify requirements of IEC 60601-1:1988 and its Amendments 1 (1991) and 2 (1995) are intended to take precedence over the corresponding general requirements.

2 Normative references

The following referenced documents are indispensable for the application of this document. For dated references, only the edition cited applies. For undated references, the latest edition of the referenced document (including any amendments) applies.

ISO 7000/IEC 60417:2004, *Graphical symbols for use on equipment — Index and synopsis*

ISO 14155-1:2003, *Clinical investigation of medical devices for human subjects — Part 1: General requirements*

ISO 14155-2:2003, *Clinical investigation of medical devices for human subjects — Part 2: Clinical investigation plans*

ISO 14937:2000, *Sterilization of health care products — General requirements for characterization of a sterilizing agent and the development, validation and routine control of a sterilization process for medical devices*

ISO 15223:2000, *Medical devices — Symbols to be used with medical device labels, labelling and information to be supplied*
Amendment 1:2002.
Amendment 2:2004.

IEC 60068-2-6:1995, *Environmental testing — Part 2-6: Tests — Test Fc. Vibration (sinusoidal)*

IEC 60068-2-27:1987, *Environmental testing — Part 2-27: Tests — Test Ea and guidance. Shock*

IEC 60068-2-32:1975, *Environmental testing — Part 2-32: Tests — Test Ed. Free fall*
Amendment 1:1982
Amendment 2:1990

IEC 60068-2-64:1993, *Environmental testing — Part 2-64: Test methods — Test Fh. Vibration, broad-band random (digital control) and guidance*

IEC 60079-4:1975, *Electrical apparatus for explosive gas atmospheres — Part 4: Method of test for ignition temperature*
Amendment 1:1995

IEC 60529:2001, *Degrees of protection provided by enclosures (IP code)*

IEC 60601-1:1988[1)], *Medical electrical equipment — Part 1: General requirements for safety*
Amendment 1:1991
Amendment 2:1995

IEC 60601-1-1:2000, *Medical electrical equipment — Part 1-1: General requirements for safety — Collateral standard: Safety requirements for medical electrical systems*

IEC 60601-1-2:2001, *Medical electrical equipment — Part 1-2: General requirements for safety — Collateral standard: Electromagnetic compatibility — Requirements and tests*

IEC 60601-1-4:1996, *Medical electrical equipment — Part 1-4: General requirements for safety — Collateral Standard: Programmable electrical medical systems*
Amendment 1:1999

IEC 60601-1-6:2004, *Medical electrical equipment — Part 1-6: General requirements for safety — Collateral standard: Usability*

IEC 60601-1-8:2003, *Medical electrical equipment — Part 1-8: General requirements for safety — Collateral standard: General requirements, tests and guidance for alarm systems in medical electrical equipment and medical electrical systems*

IEC 60825-1:2001, *Safety of laser products — Part 1: Equipment classification, requirements and user's guide*

IEC 60825-2:2000, *Safety of laser products — Part 2: Safety of optical fibre communication systems* (OFCS)

1) Currently under revision as IEC/CDV 60601-1:2004.

Attachment B – FDA (510k) Costs [20]

*Emergo Group has been helping medical device and companies with international regulatory and quality assurance issues since 1997. The company assists the manufacturer with everything from USA FDA Quality System Regulation*** (QSR) compliance and audits, to 510(k) preparation and distributor qualification.*

Shown below are the user fees the FDA charges to review 510(k) applications. An annual fee for Establishment Registration is also charged to all companies. The number in (brackets) is the discount given to "small businesses" with less than US$100,000,000 in annual sales. The fiscal year for the FDA starts on October 1 and ends September 30 each year. All prices are in USD.

FDA 510(k) Application Review Fee

Payable to the FDA to have them review a new 510(k) application. Note that the US Government fiscal year ends on September 30.

2009 - $3693 ($1847 for small businesses)
2010 - $4007 ($2004 for small businesses)
2011 - $4348 ($2174 for small businesses)
2012 - $4717 ($2359 for small businesses)

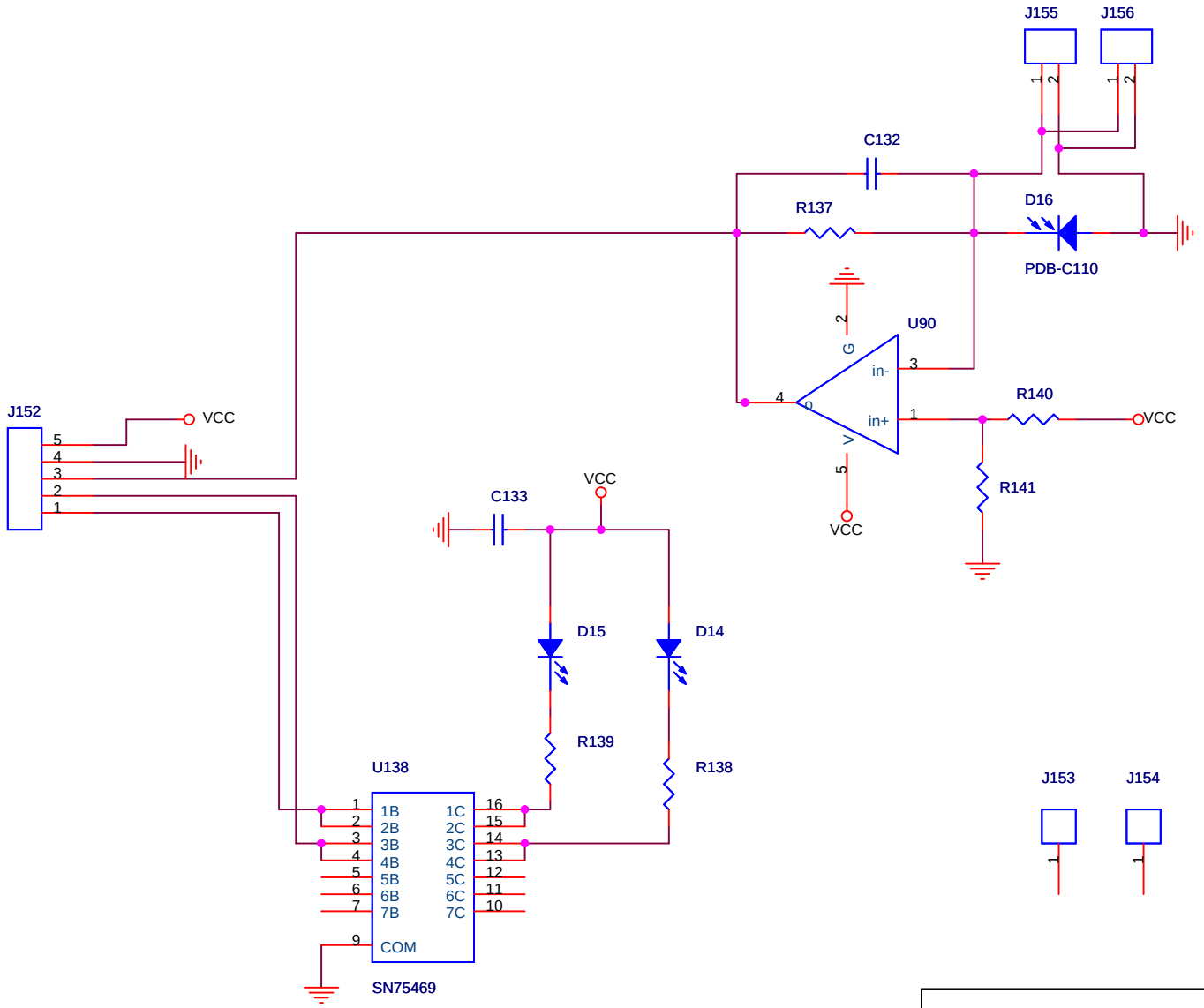
FDA Establishment Registration Fee

Payable once per year by every registered medical device company. No discount is provided to small businesses for the FDA Establishment Registration fee.

2009 - $1851
2010 - $2008
2011 - $2179
2012 - $2364

Attachment C

The first oximeter probe circuit (September, 2008)



Title	Oxymeter	
Size A4	Document Number	Rev 1
Date: Tuesday, September 02, 2008	Sheet 7 of 12	

Attachment D – Matlab Algorithm

```
close all;
clear all;
clc

% Open the File Text

num_data = 12000;
fid = fopen('exemplo.dat','r');

for i=1:10
    tline = fgets(fid);
    i=i+1;
end

% Choice of the Range

lower = 1;
upper = 8000;

% Convert Values

i = 1;
data(num_data,2) = zeros();
while(i <= num_data)
    data_stream = fgets(fid);
    cdata_stream = regexprep(data_stream, ',', '.');
    a = sscanf(cdata_stream,'%f %f');
    data(i,1) = a(1);
    data(i,2) = a(2);
    i=i+1;
end

% Choice of the Variables

x = data(lower:upper,1);
y = data(lower:upper,2);


% Smooth Function

z1=smooth(y,10);

% Peaks Detection

[maxtab, mintab] = peakdet(y', 0.06, x')

[smaxtab, smintab] = peakdet(z1', 0.015, x')

% Graphical Representation of the Signals

figure;

subplot (2,2,1)
plot(x,y)
xlabel('Time (sec)')
```

```
ylabel('Voltage (V)')

subplot (2,2,2)
plot(x,z1)
xlabel('Time (sec)')
ylabel('Voltage (V)')

subplot (2,2,3)
plot(x,y,'b-',...
mintab(:,1), mintab(:,2), 'g*',...
maxtab(:,1), maxtab(:,2), 'r*');
xlabel('Time (sec)')
ylabel('Voltage (V)')

subplot (2,2,4)
plot(x,z1,...
smintab(:,1), smintab(:,2), 'g*',...
smaxtab(:,1), smaxtab(:,2), 'r*');
xlabel('Time (sec)')
ylabel('Voltage (V)')


fclose(fid);


% Heart Rate

heartbeat1= length (maxtab)*6

heartbeat2= length (smaxtab)*6
```

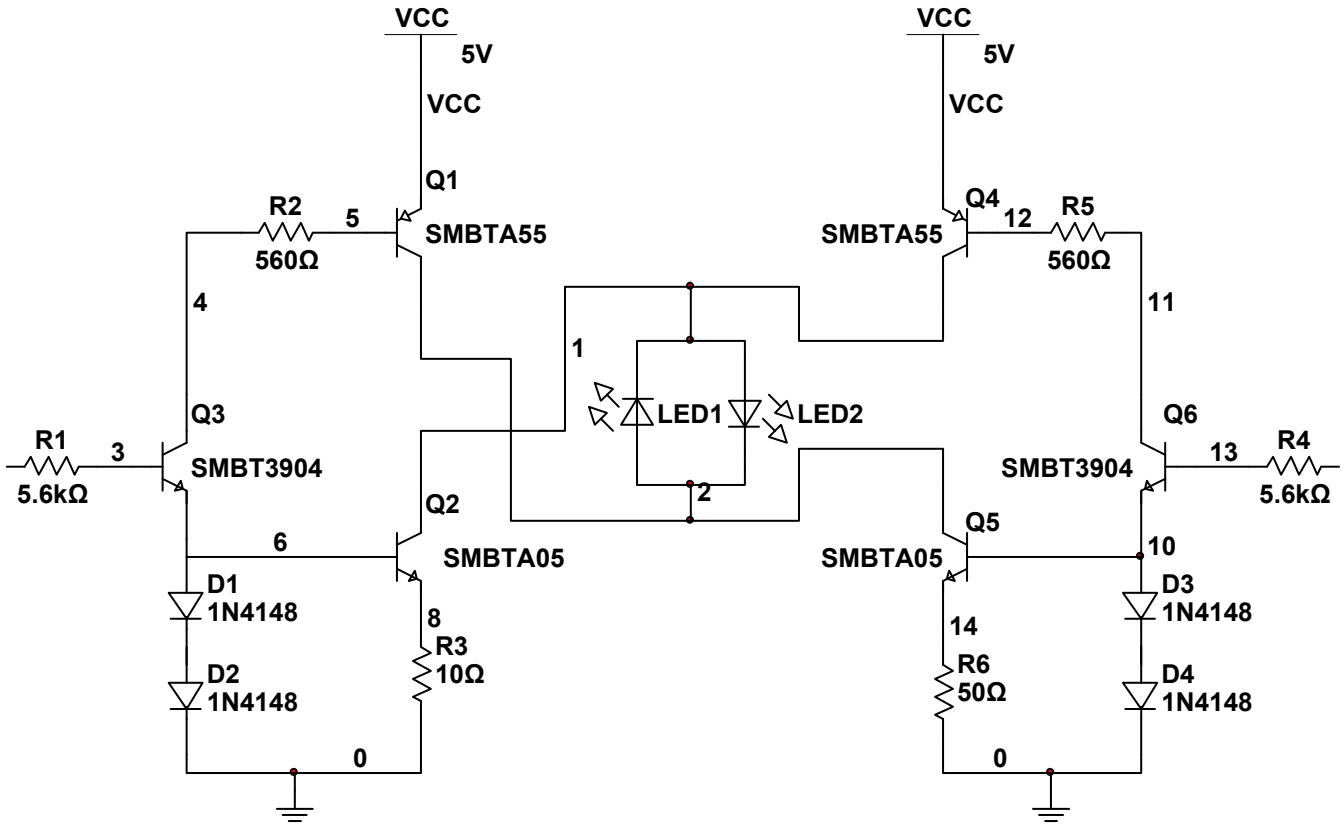
Attachment D – Peakdet Algorithm

```
function [maxtab, mintab]=peakdet(v, delta, x)
%PEAKDET Detect peaks in a vector
%        [MAXTAB, MINTAB] = PEAKDET(V, DELTA) finds the local
%        maxima and minima ("peaks") in the vector V.
%        MAXTAB and MINTAB consists of two columns. Column 1
%        contains indices in V, and column 2 the found values.
%
%        With [MAXTAB, MINTAB] = PEAKDET(V, DELTA, X) the indices
%        in MAXTAB and MINTAB are replaced with the corresponding
%        X-values.
%
%        A point is considered a maximum peak if it has the maximal
%        value, and was preceded (to the left) by a value lower by
%        DELTA.

% Eli Billauer, 3.4.05 (Explicitly not copyrighted).
% This function is released to the public domain; Any use is allowed.

maxtab = [];
mintab = [];

v = v(:); % Just in case this wasn't a proper vector

if nargin < 3
  x = (1:length(v))';
else
  x = x(:);
  if length(v)~= length(x)
    error('Input vectors v and x must have same length');
  end
end

if (length(delta(:)))>1
  error('Input argument DELTA must be a scalar');
end

if delta <= 0
  error('Input argument DELTA must be positive');
end

mn = Inf; mx = -Inf;
mnpos = NaN; mxpos = NaN;

lookformax = 1;

for i=1:length(v)
  this = v(i);
  if this > mx, mx = this; mxpos = x(i); end
  if this < mn, mn = this; mnpos = x(i); end

  if lookformax
    if this < mx-delta
      maxtab = [maxtab ; mxpos mx];
      mn = this; mnpos = x(i);
      lookformax = 0;
    end
```

```
  else
    if this > mn+delta
      mintab = [mintab ; mnpos mn];
      mx = this; mxpos = x(i);
      lookformax = 1;
    end
  end
end
```

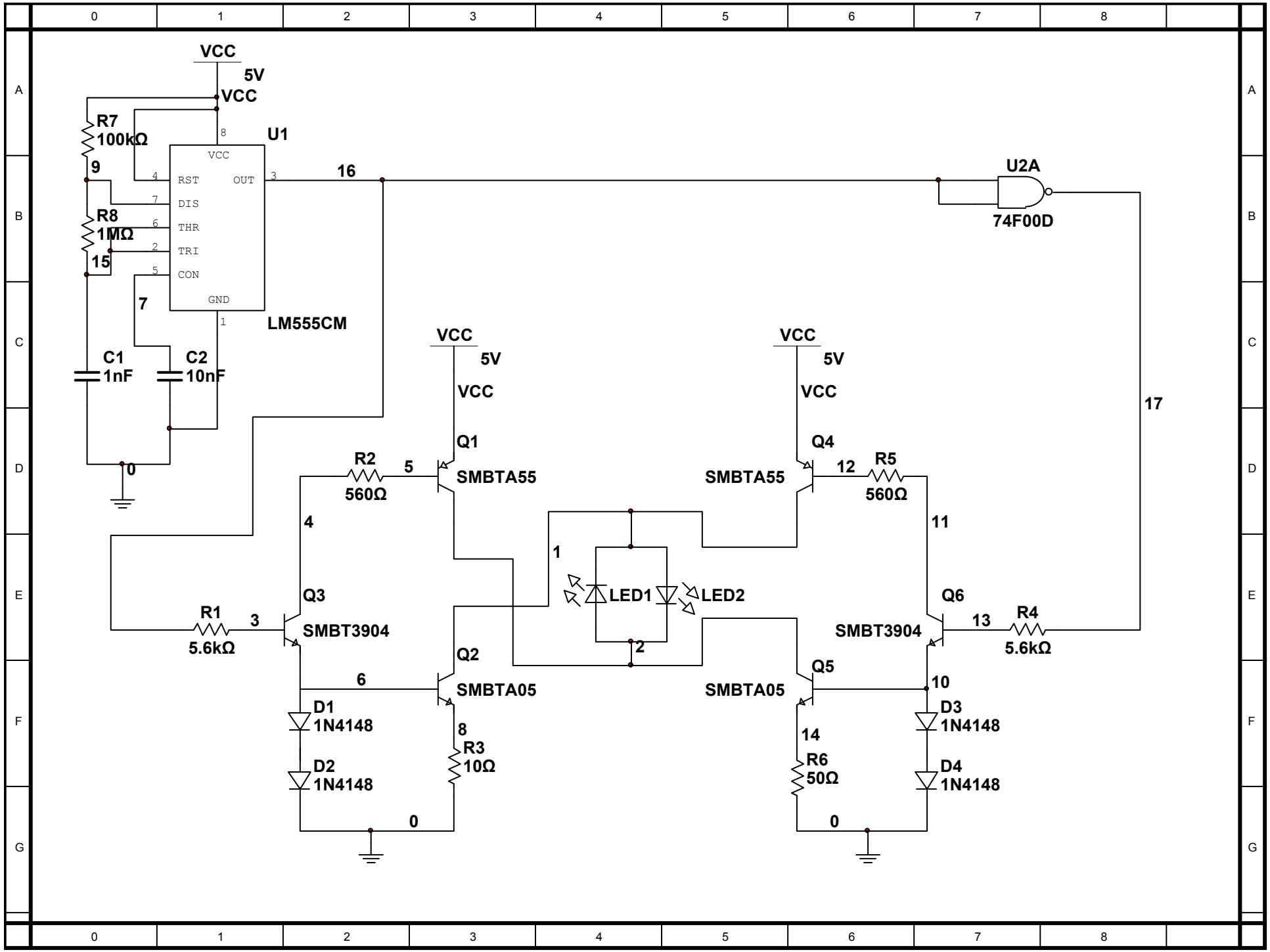
Attachment F

LED's Driver Circuit


VCC
5V
VCC
Q1
R2
5
560Ω
SMBTA55
4
R1
3
5.6kΩ
Q3
SMBT3904
1
LED1
LED2
2
Q2
6
SMBTA05
D1
1N4148
D2
1N4148
8
R3
10Ω
0
VCC
5V
VCC
Q4
12
R5
560Ω
SMBTA55
11
Q6
13
R4
5.6kΩ
SMBT3904
Q5
10
SMBTA05
D3
1N4148
D4
1N4148
14
R6
50Ω
0

Attachment G

LED's Driver Circuit (including timing)


VCC
5V
VCC
R7
100kΩ
9
8
U1
VCC
RST
OUT
DIS
THR
TRI
CON
GND
LM555CM
R8
1MΩ
15
7
1
C1
1nF
C2
10nF
0
16
U2A
74F00D
17
VCC
5V
VCC
Q1
R2
560Ω
5
SMBTA55
4
R1
5.6kΩ
3
Q3
SMBT3904
1
LED1
LED2
2
Q2
6
SMBTA05
D1
1N4148
D2
1N4148
8
R3
10Ω
0
VCC
5V
VCC
Q4
SMBTA55
12
R5
560Ω
11
Q6
SMBT3904
13
R4
5.6kΩ
Q5
SMBTA05
10
D3
1N4148
D4
1N4148
14
R6
50Ω
0

Attachment H – 555 Astable Frequencies [28]

555 astable frequencies			
C1	**R2 = 10k R1 = 1k**	**R2 = 100k R1 = 10k**	**R2 = 1M R1 = 100k**
0.001µF	68kHz	6.8kHz	680Hz
0.01µF	6.8kHz	680Hz	68Hz
0.1µF	680Hz	68Hz	6.8Hz
1µF	68Hz	6.8Hz	0.68Hz
10µF	6.8Hz	0.68Hz (41 per min.)	0.068Hz (4 per min.)