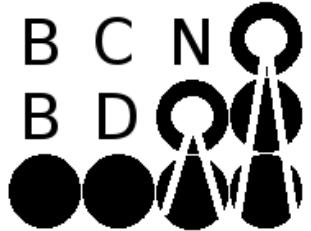


Barcelona Biomedical Devices



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Design of an Electrocardiography and Respiration Device for Ambulatory Monitoring

Joan Muñoz Gayet

Final Report

1. ABSTRACT

The objective of this project is to create a device able to monitor at the same time the respiration and the electrocardiogram (ECG) of a patient in ambulatory situations. From these signals, several data can be extracted in order to avoid possible dangerous situations or detect cardio-respiratory pathologies. The final product has to be small and portable in order to be easily worn in any context and effortlessly interfaced with the user. For this reason, the connection to the patient is simplified to only two electrodes at the two sides of the thorax, the device size is approximately the one of a smart phone and the communication is performed by *Bluetooth* in order to be easily extendable to mobile phones applications or other future implementations.

2. ACKNOWLEDGEMENTS

I would like to express my best thanks to my project supervisors, Mireya Fernandez and Juan Ramos, who have always helped me in every single moment I have had problems.

The cooperation of the laboratory professors, Alfonso Méndez, Vicente Ruiz and Manel Garcia, has also been indispensable in my project. They all perform an essential work in this university.

To all teachers from this university who have taught me how to become an engineer.

To my parents who have always supported me under any situation and inspired me to become the person I am.

To my grandparents whose wisdom and experience have taught me how to live in this world.

To all my friends who have stolen a smile from me in my worst moments.



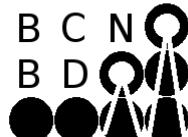
3. REVISION HISTORY AND APPROVAL RECORD

Revision	Date	Purpose
0	15/11/2013	Document creation
1	26/11/2013	Document revision

DOCUMENT DISTRIBUTION LIST

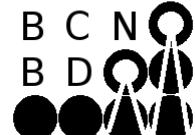
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5. TIME PLAN UPDATED

The time plan from the critical review has been strictly followed. The only modification produced has been in the last tasks: a Printed Circuit Board (PCB) has been made in order to improve the performance. This has delayed the final tasks one week but has not become a problem as there was a spare week before the dead time of the project. Basically, tests have been one week delayed as the following Gantt diagram states.

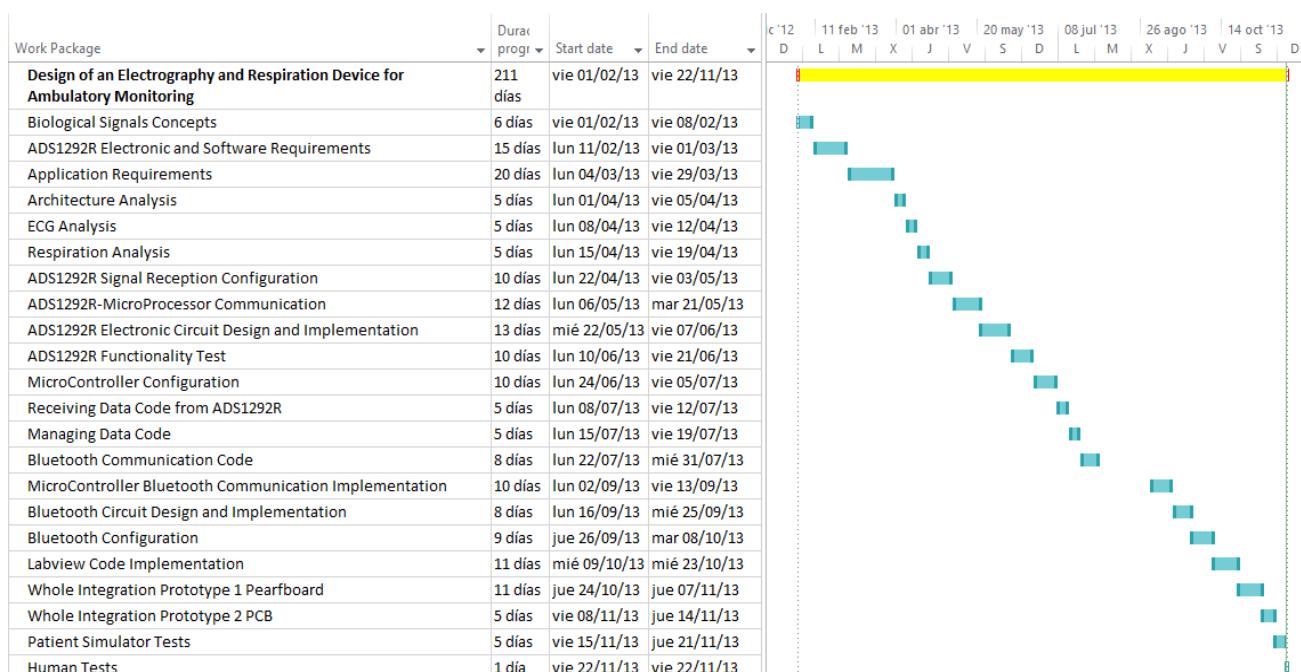
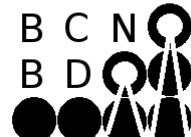


Figure 1: Gantt diagram



6. INTRODUCTION

The main objective of this project is to monitor the respiration and electrocardiographic (ECG) signals at the same time in ambulatory situations. Despite the fact that there are already components and devices that provide these functionalities in clinical situations, there is not anyone that performs both things in a portable integrated chip that can be worn in any context. These signals could be used for many applications and to diagnose multiple pathologies as cardio-respiratory problems.

There are several research projects which are working in the same concept using different technologies. For example, in the Electronic and Biomedical Instrumentation Group of the UPC, collaborating with the company *Health and SportLab*, it is being investigated that, from the ECG signal, the psychological and physical states of a patient can be estimated. Then, the stress level or the state of health of a patient are studied in relation to the physic exercise practiced and the nutrition model that is carried out. And basically, this is implemented only by using a thoracic band that includes a small module which measures the heart rate and transmits it wiressly.

Other simultaneous research projects in the same department are trying to measure the respiration of a person during his/her daily activity. To achieve this objective multiple technologies are being used:

- Using an inductive band connected around the thorax, the respiration can be extracted. Depending on its deformation, the inductive band size is incremented and also its inductance value. Monitoring the inductance changes, the inspiration and exhalation of a patient can be measured [1].
- Using textile capacitive sensors connected on the driver's seat, belt or around the steering wheel, the respiration rate can be recorded [2].
- Using cameras which detect the chest movements, the air flux or the eyes closure, the patient's respiration can also be estimated [3].

The main objective of all these projects are to detect drowsiness or levels of fatigue that might become dangerous when someone is driving or making things that need full attention. The objective is clear: detect when somebody is falling asleep.

As a result of this interest, several companies are offering products that implement one of these functionalities. For example, the company *FaceLake* [4] is manufacturing a real time portable ECG monitor. Furthermore, the company *alivercor* [5] is selling a product that, from two electrodes allocated in the two sides of the device, can provide the ECG signal just with the contact of the fingers.

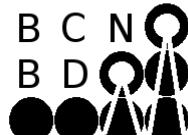
Moreover, multiple electronic devices are already designed to provide bioimpedance analysis such as the AD5933 [6]. These components can be used to detect, for example, the Total Body Water (TBW) or as the case of this project, the respiration signal. Actually, the bioimpedance analysis has become more and more popular during the last decades for medical purposes [7] [8].

When the patient is falling asleep, his respiration becomes more irregular, slower and deeper. A detector of this pattern in drivers would save many lives. Other pathologies like insomnia or snoring might be also detected and in case of risk, the emergency services would be called. Actually, this goal is not new at all and several studies have been made to detect drowsiness in patients [9]. However, the objective of this work is not analyzing or studying how to detect these patterns but to make up a prototype which provides this signal.

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Multiple future applications can be conceived from a device that provides both respiration and ECG signals: sudden cardiac death detector, arrhythmia monitor, airway obstruction alarm, respiratory diseases diagnosis... The only problem is to develop something as compact and tiny as possible in order to be easily implemented or connected to anybody. The price is another achievement to overcome. In case a small and cheap module might be manufactured, the commercialization might become easily extendable. Lorry drivers could carry this product compulsorily or high risk people shall wear it in order to avoid or detect dangerous situations. Sportsmen and sportswomen could wear it in order to monitor improvements in their physical condition.



7. PROJECT BACKGROUND

a. Electrocardiography

The definition of the electrocardiogram (ECG) is the electrical activity of the heart. The signal represented in Figure 2 is the typical cardiac cycle (heartbeat). The baseline is 0 mV, the frequency is normally between 60 and 100 bpm and the voltages are between 0.1 and 10 mV. The frequency spectrum is from 0.05 to 150 Hz [9].

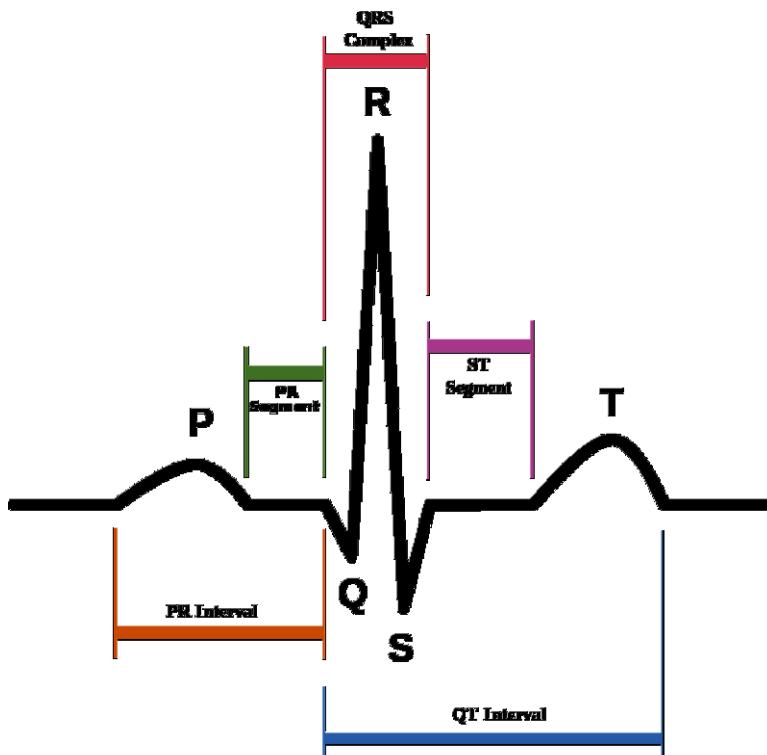


Figure 2: Typical cardiac cycle [10]

The first wave, the P wave, is caused by the atrial depolarization (prior to contraction). Then, there is the QRS complex produced by the ventricles. This second wave is bigger as the ventricle muscle mass and pressure are higher. Finally, T wave is the representation of the repolarization of the ventricles. In some patients, it also appears another very small wave named U-wave but its origins is not clear at all and multiple hypotheses are being analyzed [11].

To obtain this information, there are different connections to the patient called ECG leads. Some combinations of different connections have been standardized and known as standard leads. Usually ten electrodes are used to obtain twelve different combinations (leads). Most common and important leads are I, II and III and they use electrodes connected to the Right Arm (RA), Right Leg (RL), Left Arm (LA) and Left Leg (LL). The one that is used in this project is Lead I which has got the positive input connected to the LA and the negative one to the RA [9].

The electrical meaning of the three first leads is shown in the Einthoven triangle diagram (Figure 3) where the polarity along the RA, LA, RL and LL surrounding the heart is shown. Basically, the Einthoven's law states that the sum of the voltages in any 2 ECG leads equals the voltage in the remaining lead [9].

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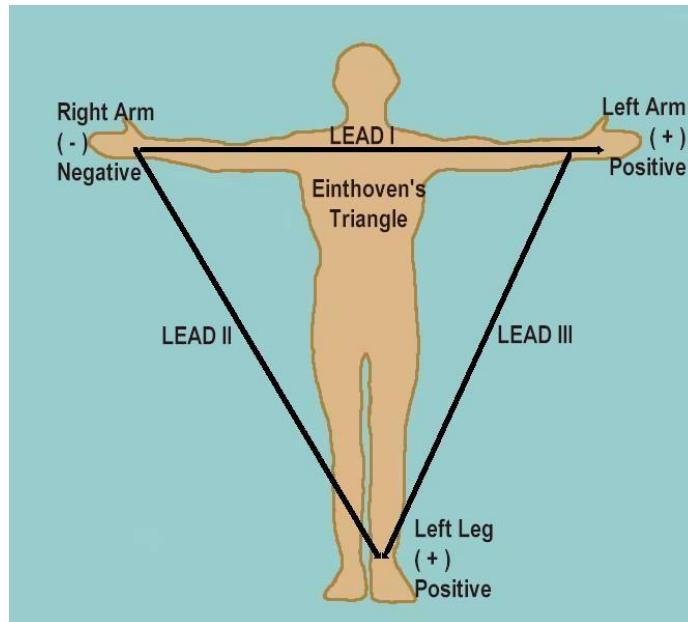
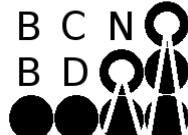


Figure 3: Einthoven's triangle [12]

There is lot of information that can be extracted from the ECG signal: the heart rate, detection of abnormalities in the propagation of the impulse, detection of coronary diseases or heart attacks... So, extracting at least one lead might provide essential information.

b. Respiration

The other signal to be measured in this project is the respiration. So, firstly, a deep analysis of its characteristics has to be performed in order to properly design the hardware and software.

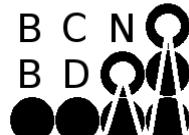
The normal or common respiration is defined by the Tidal or Total Volume (TV) which is the amount of air flowing into and out of the lungs which is more or less 500 ml in an adult and starts from the Resting Expiratory Level (REL) [13].

In spite of not being the purpose of this project, there are other important air volumes that might provide important data and do not come from the normal respiration features but from the anatomic characteristics (Figure 3).

The Inspiratory Reserve Volume (IRV) is defined by the ability to inspire more than the TV. If the Tidal Volume is added, the Inspiratory Capacity (IC) is got which is approximately seven times the TV.

On the other hand, the maximum exhalation from REL is the Expiratory Reserve Volume (ERV). However, there is still air in the lungs when all the ERV is extracted: the Residual Volume (RV). But this last volume cannot be exhaled. The addition of ERV and RV is the Functional Residual Capacity (FRC).

There are many dynamic tests based on the respiration signal: Forced Expiratory Volume, Maximum Voluntary Ventilation, Forced Vital Capacity... And, actually, there are many methods to



acquire these data: using spirometers, oxygen uptakes... But in this project, the tidal volume is obtained by electrical bioimpedance.

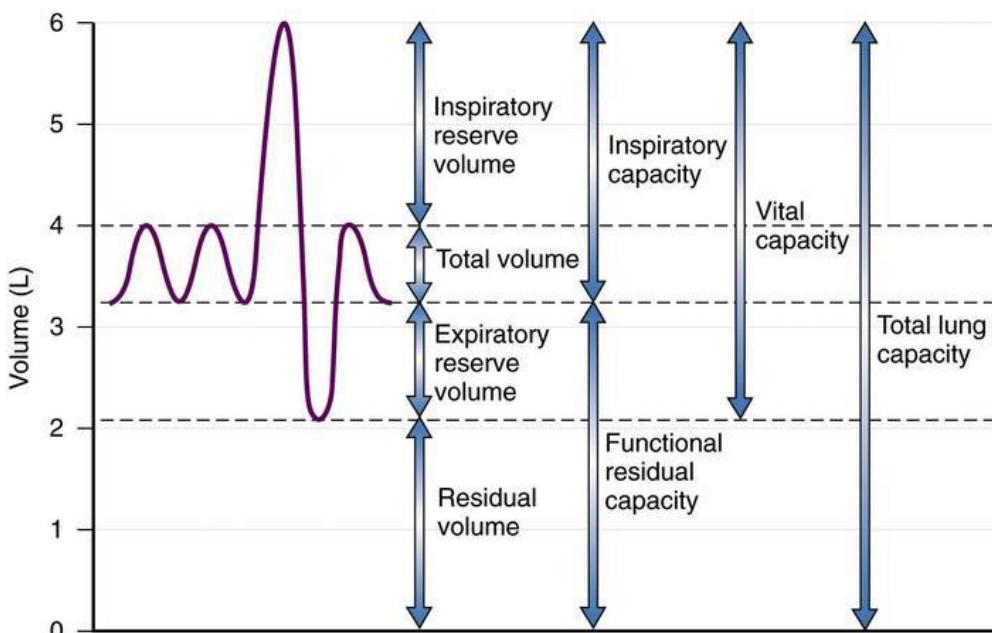


Figure 4: Lung air volumes [14]

c. Electrical Bioimpedance

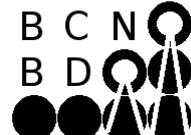
The Electrical Bioimpedance (BEI) has been used since the turn of XX century. Using a frequency excitation between 50 and 100 kHz the equivalent impedance of an object can be obtained. This method is commonly used for several purposes: apnea monitoring, detection of venous thrombus, peripheral blood flow, body composition... [13].

In this project, the impedance to be measured is the patient's thorax. Depending on the modification of the volume of air in the lungs, the equivalent impedance changes ($1-2 \Omega/l$ of lungs volume) and, consequently, the respiration signal can be extracted. The best place to connect the electrodes is close to the armpit as the obtained signal is big enough and lineal [13].

To perform a BEI there are two implementations: using two or four electrodes techniques. The second case provides more accurate measurements as the contact equivalent resistance does not introduce measurement errors. But, for this product, it is not needed an exact equivalent impedance value but its change along the time. So, there is no reason to use the four electrodes method as well as using the two electrodes technique allows to the final application become more compact and comfortable.

d. Project Requirements

The developed product has to be able to measure the electrocardiographic (ECG) and respiration signals. Both magnitudes have to be processed, filtered and finally presented to the final user. In this project, all the captured information is going to be presented in a computer screen using a



human machine interface built in *Labview* program. But this does not mean that improved applications might display this data into a smart phone. This product should provide a start point of multiple possible applications such as a detector of car driver's drowsiness, infarction or other possible pathologies.

Obviously, this should be autonomous and portable. So, it is going to be fed from AAA batteries with a capacity of 1000 mAh.

The final prototype has to be small, light, comfortable and easy to use for the patient.

As a conclusion, the objective of this project is to develop a final user application able to perform the basics goals: show to the patient his ECG and respiration signals.

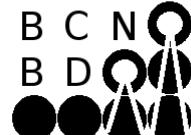
e. Project Specifications

As stated before, this product has to be portable and also small. The final size is approximately the one of a smart phone (105 x 58 x 18.5 mm).

The batteries capacity has to provide an autonomy of, at least, ten hours and, using AAA batteries, this can be achieved.

There are multiple electrical signal considerations that should be taken into account:

- The ECG frequency spectrum is between 0.05 and 150 Hz. However, in this application, the 0.05 to 40 Hz range is the only part considered. This is because it provides the basic information that is currently needed. Future achievements might also use bigger frequency spectrum ranges.
- The input signal range is between ± 300 mV.
- The sampling frequency has to be high enough to not have *aliasing* effects and to allow eliminating as high frequencies as possible. A sampling frequency of 500 Hz is enough.
- The CMRR is not critical as it is battery powered.
- Noise should not be higher than 10 μ V. ADS1292R specifications state that the SNR should be 105 dB and the CMRR -120 dB at 3V of supply and a data output rate of 500 samples per second.
- The *Bluetooth* communication uses version 2.0.



8. SYSTEM DESIGN

a. Block diagram

The project architecture is based in four different blocks:

- The ADS1292R electronic circuit
- The microcontroller
- The *Bluetooth*
- The *Labview*

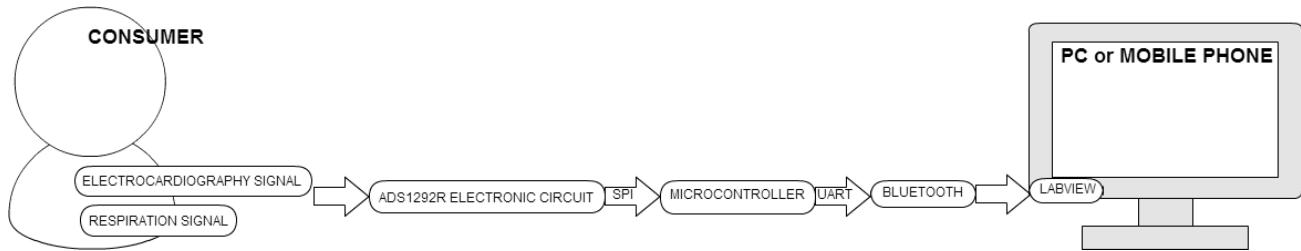


Figure 5: General architecture design

The first block, the ADS1292R electronic circuit, measures the respiration and electrocardiographic signals and send them using a Serial Port Interface (SPI). It is absolutely important that the power supply of this part has got as less noise as possible as the voltages to be measured are lower than 1mV. Other applications could measure other ECG derivations instead of the respiration. Nevertheless, the global structure would be exactly the same.

Secondly, the microcontroller receives all these data, change its format from Ca2 to decimal and, finally, sends it using a UART port. The configuration of the ADS1292R has to be also performed by this module. This means that the code has got two different parts: the configuration and the communication. The second part has to be fast, at least, it has to be executed once every 2 ms (a 500 Hz frequency) in order to not have *aliasing* effects in the measured signals and to filter as much noise as possible.

The *Bluetooth* module receives the data from a UART port and resends it using a *Bluetooth* antenna. Actually, this module is transparent; it just changes a UART communication to *Bluetooth* (Serial Port Profile).

And finally, the *Labview* program gets the transmitted data and plots both ECG and respiration signals. Apart from these representations, other important and implicit parameters could be shown: the cardiac and respiration rhythms, the noise...

b. Evaluation Board

Before the first prototype design started, some tests with the ADS1292R Demonstration Kit were performed. This evaluation board is able to capture the ECG and respiration signals and send them to the PC using a RS-232 cable. It is provided by the Texas Instruments company in order to show

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that their product ADS1292R works properly. So checking its functionalities is essential for this project as it is the main component in the general architecture.

It is good to notice that despite the fact that this demonstration kit provides the basic functionalities that this project performs, there are other goals and features that the final product offers:

- Smaller package
- Simpler microprocessor
- *Bluetooth* communication
- Electrical protections for the patient and for the product
- Less energy consumption

So, some tests were done in order to analyze whether this product suits this project or not. To do that, three electrodes were connected to a volunteer: two to the left and right chest sides to get the differential signal and another to right part of the stomach to connect the reference.

i. Respiration Tests

The respiration obtained from the evaluation board is good enough. Figure 6 shows an example of the signal captured by the ADS1292R Demonstration Kit. Both inspiration (maximum) and exhalation (minimum) phases can be easily been identified; in this example, there are eight inspirations and exhalations.

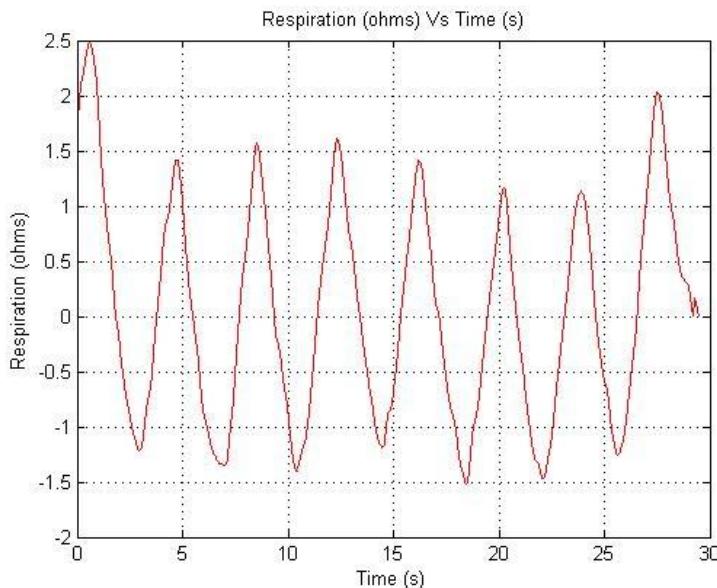


Figure 6: Respiration signal obtained from the ADS1292R Demonstration Kit

ii. Verification Tests

The example shown before seems to be totally fine. However, it would be a good idea to compare the signal obtained with the one using other technologies. An inductive band has been employed in order to perform this comparison. This technology measures the deformation of a band that is attached around the chest depending on the volume of air inside the lungs: when the patient is inhaling, the band becomes bigger, on the other hand, when the patient exhales, the band becomes smaller. The change in its size produces change in the inductance that is measured. Figure 7 shows the comparison using these two technologies.

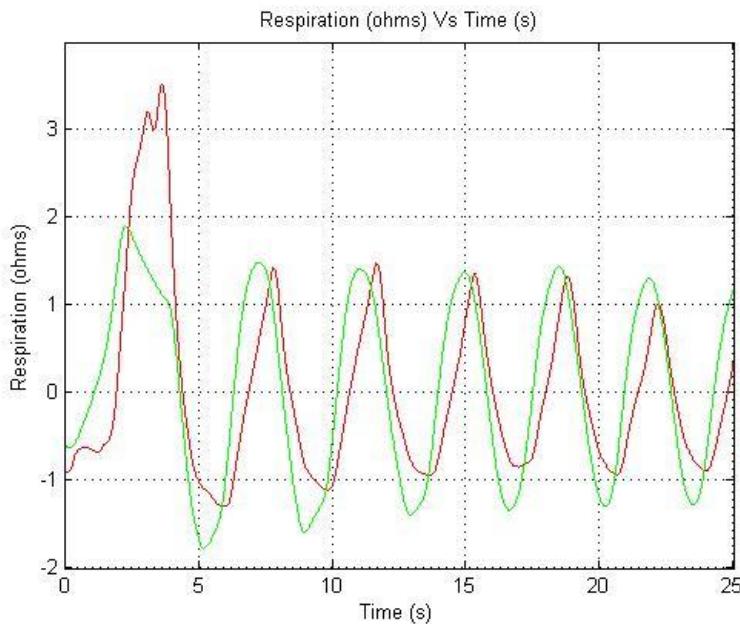
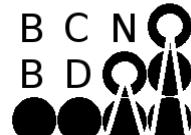


Figure 7: Comparison of the respiration signals obtained by the ADS1292R Demonstration Kit (red) and the inductive band (green)

Both signals have been normalized by its standard deviation in order to have approximately the same amplitude. So, the Y-axis units are actually not ohms: it is true that the ADS1292R measures the real part of impedance but it is not the case of the inductance band. However, as the objective is to compare both shapes, this is not a problem.

It is important to notice that to compare both signals a synchronization process had to be performed in order to be in the same phase of the respiration in both signals. This was done by iterating the position of one signal respect to the other. The position where the error is minimum should be the synchronized place. However, this is not actually true (as it can be observed in Figure 7) and both signals have a little time mismatch. This is because these signals come from different technologies. But both graphics have got more or less the same shape. So, consequently, it can be concluded that the method using the ADS1292R Demonstration Kit works properly.

iii. Artefacts detection

As it has been said before, one of the main objectives of this project is to record the respiration signal. From it, other information can also be extracted: some artefacts could be detected using this technology. An artefact happens when the diaphragmatic or intercostal muscles change its position [13]. It can be the movement that a driver does when he sneezes or the movement when he moves the steering wheel.

To demonstrate this fact, the respiration signal in two different situations, having and not having artefacts, has been captured. Figure 8 shows the respiration signal of a volunteer just sat down. And Figure 9 is the respiration of the same volunteer but moving a steering wheel first to the right, then to left two times and, finally, to right another time.

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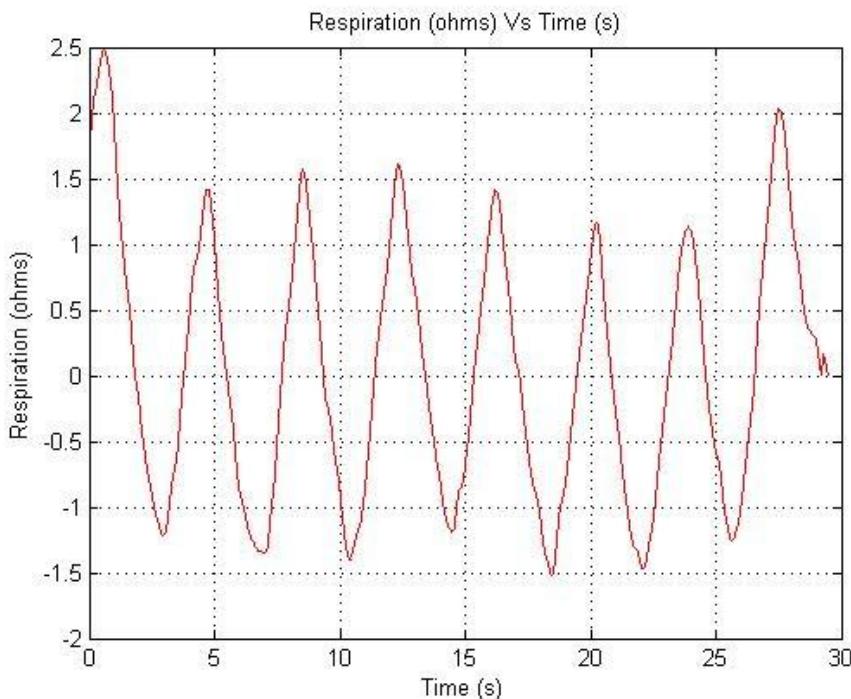


Figure 8: Respiration signal of a volunteer without artefacts

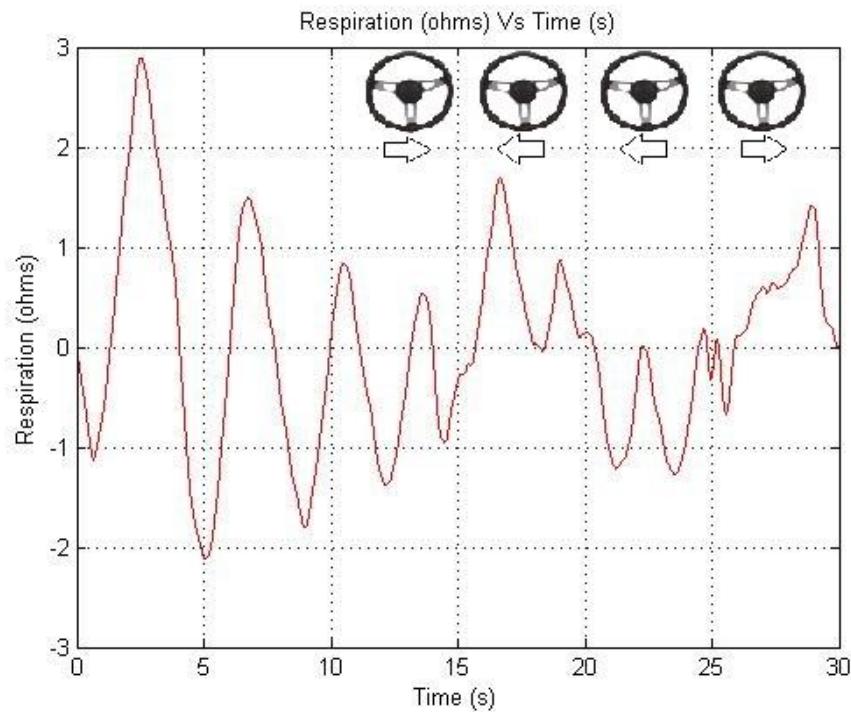
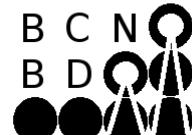


Figure 9: Respiration signal of a volunteer with artefacts

As it can be observed from Figure 9, it is easy to realize that something is happening to the volunteer but not what. However, as a simple speculation, getting the pattern of every movement, the artefact could be identified. In this case, for example, the movements of the volunteer were previously known and, consequently, they can be identified. At the 12th second, the volunteer starts moving the steering wheel to the right and the respiratory signal goes up a little. Then, when the

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volunteer moves the steering wheel to the left two times, the signals goes down twice and making some strange patterns. Finally, when the volunteer moves the steering wheel to right another time, the signal goes up and, in this case, there is also some strange patterns.

iv. ECG Tests

Apart from the respiration signal, this product also detects the electrocardiogram. From it, much information can also be extracted. In Figure 10, an example of an ECG is shown.

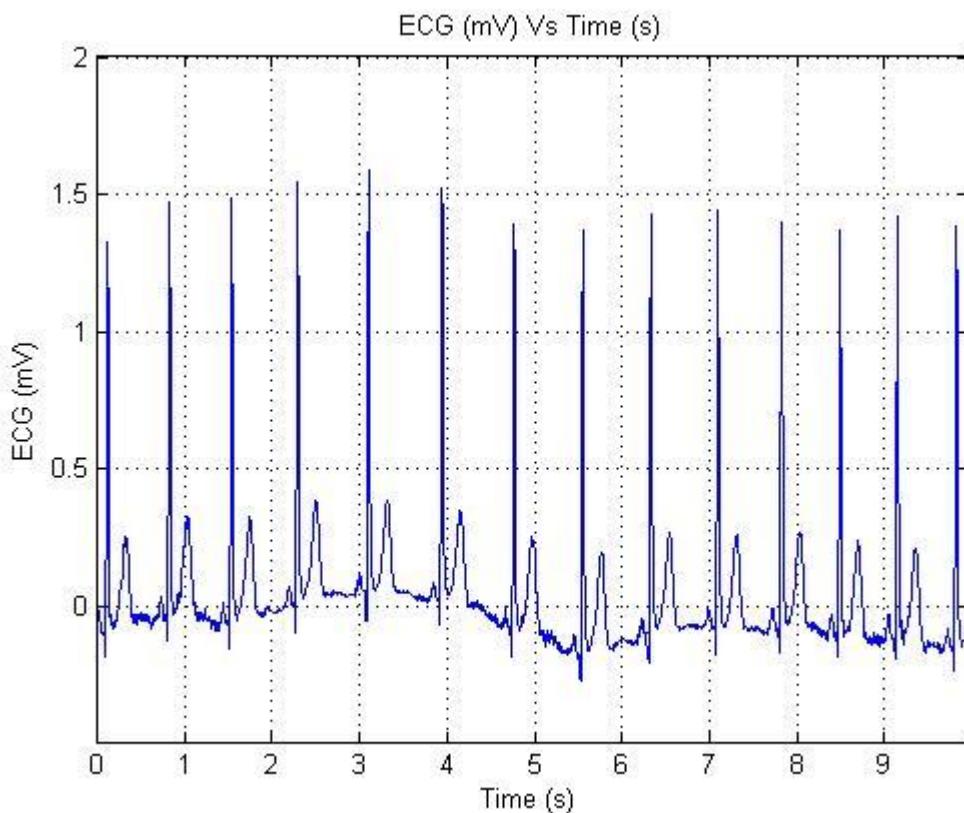


Figure 10: ECG signal of a volunteer

Obviously, this signal should be filtered in order to avoid interferences but, in this case, it has not been implemented in order to show other type of inputs. For example, a very low frequency is visible: it corresponds to the respiratory signal. So, if the output were filtered in different band pass filters, both signals, ECG and respiration, might be recorded. Furthermore, the interference of 50 Hz is also visible. Figure 11 shows the FFT of the ECG signal.

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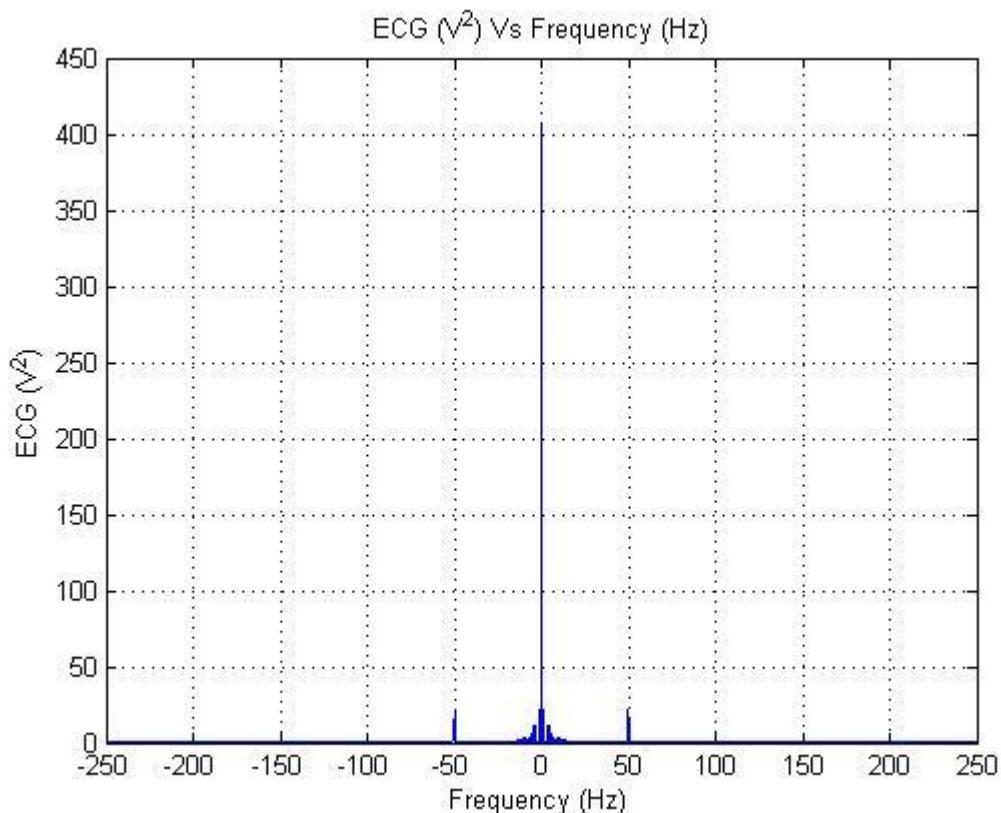
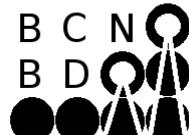


Figure 11: FFT of the ECG signal showing the 50 Hz interference

c. ADS1292R

The most important component in the general architecture is the ADS1292R. This element manufactured by the Texas Instruments company is able to sample the ECG and respiration signals, convert them from analog to digital and send this information using Serial Port Interface (SPI) (Figure 12 shows the general internal architecture).

These are the functionalities this project uses to achieve its goals but the ADS1292R can sense other signals:

- Instead of recording the respiration, it could also measure other leads.
- Lead off detection. This functionality should be used in future prototypes but it is not currently implemented.
- Temperature measurement from a sensor implemented on-chip.
- Test signal. This functionality is not used either but future improvements of the final application could use it in order to calibrate the circuit.

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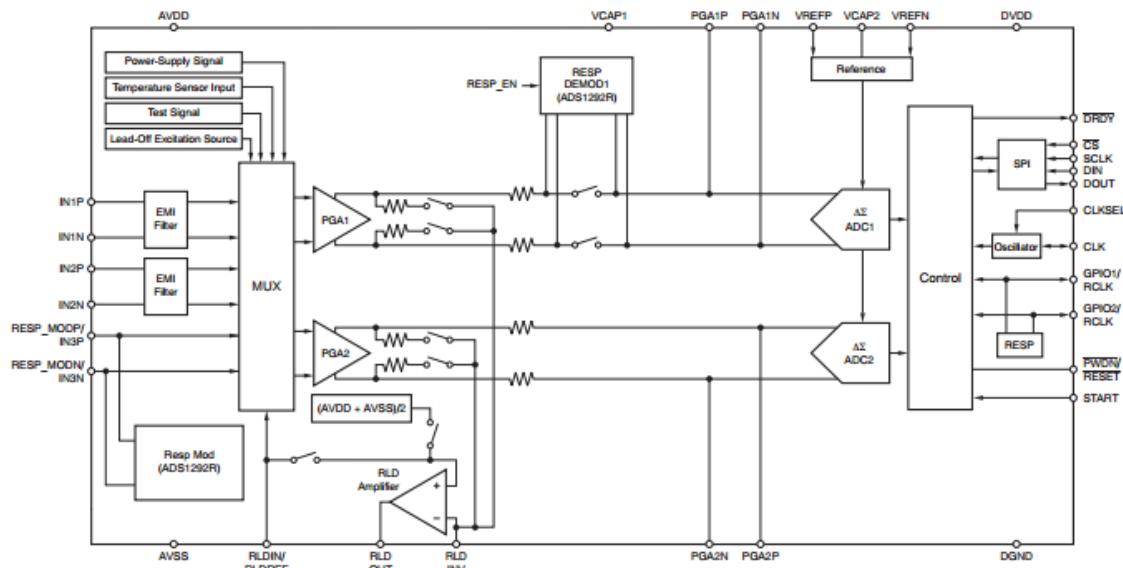
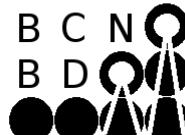


Figure 12: General internal architecture

i. Electrical Characteristics

The most important electrical characteristics of the ADS1292R are stated in Table 1.

PARAMETER	VALUE
DC Input Resistance	1000 MΩ (minimum)
Bandwidth	8.5 kHz (typical)
Analog to Digital Converters (ADC) Resolution	24 bits
Date Rate	8000 (maximum)
Common Mode Rejection Ratio (CMRR)	-105 dB (minimum)
Power Supply Rejection Ratio (PSRR)	90 (typical)
Total Harmonic Distortion (THD)	-82 dB (typical)
Analog Supply (AVDD)	2.7 – 5.25 V
Digital Supply (DVDD)	1.7 – 3.6 V
Total Current Consumption	325 µA (typical)

Table 1: Most important ADS1292R electrical characteristics [15]

ii. Schematic

It is important to understand that the ADS1292R performs an electrical bioimpedance by generating a square signal (from RESP_P and RESP_N pins) connected to the patient that is converted to a current using two series resistors. By this way, the equivalent real impedance can be obtained measuring the voltage across the patient chest (electrodes). This impedance could be sensed using two or four wires methods. But, as in this project the main objective is to record the respiration cycles and the pragmatism of the final product is absolutely important to achieve, the two electrodes method is the one used and two electrodes are being used to inject current and to measure the voltage drop.

Other important considerations are the filters that must be implemented in each input to reject the undesired frequencies. IN1N and IN1P are the differential inputs for the respiration signal



measurement. As the current excitation is in AC (32 kHz), it is not wanted to get voltages at low frequencies nor at DC. For that reason, two series capacitors are included to filter those frequencies. On the other hand, IN2N and IN2P are the differential inputs for the ECG signal measurement and, as this signal is in low frequencies, a low pass filter is implemented (R-C network).

Other important parameters, such as the Right Leg Drive (RLD) that compensates the common mode signal or the bypassing capacitors, are exactly the same that Texas Instruments recommends.

In terms of supply, both digital and analog supplies have the same voltage (3.3 V) but they are separated by a ferrite in order to avoid interferences in the analog part of the circuit.

Despite the fact that the manufacturer does not recommend any safety protection, some precautions have been taken into account. For example, in series to the patient, a 10 k Ω series resistance has been added to limit the current through the patient.

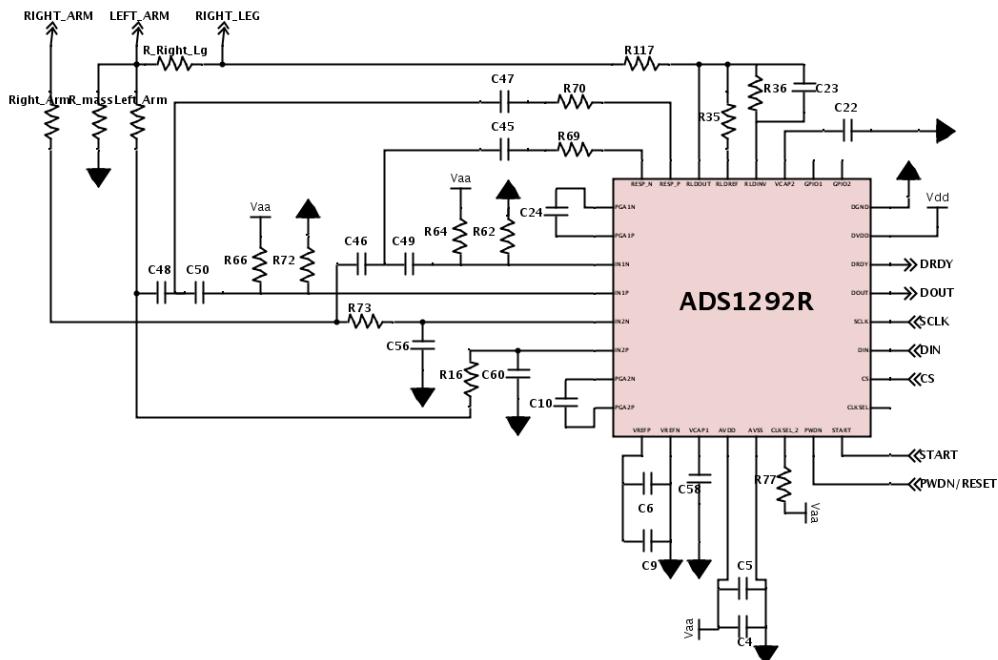
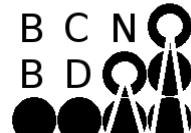


Figure 13: ADS1292R schematic

d. Microprocessor Circuit Design

The microprocessor used in the ADS1292R Demonstration Kit [16] is the MSP430F5529IPNR [17]. However, this micro is manufactured by the Texas Instruments company and there are not available programmers in the lab to program it. Basically, this component should be able to communicate to the ADS1292R by SPI, perform a little data processing and send it using a UART port. So, as the available programmer was from the Microchip (Pickit 3 [18]) company and there are several micros that fulfill the needed specifications, a microprocessor from this company was selected: PIC18f24j11 [19].



i. Characteristics

PIC18f24j11 microchip has got the following features (Table 2):

PARAMETERS	VALUE
Low Power Technology	NanoWatt XLP
Supply Voltage	2 – 3.6 V
2 Master Synchronous Serial Ports	SPI, I2C
2 USART modules	RS-485, RS-232, LIN/J2602
External Clock Configuration	Up to 48 MHz
I/O PINS	18
Programmer	MPLAB PM3, MPLAB REAL ICE, MPLAB ICD2, MPLAB ICD3, Pickit2, Pickit 3
Packages	28-SPDIP, 28-SOIC, 28-SSOP , 28-QFN, 44-QFN, 44-TQFP Package
Program Memory	16 kBytes
Data Memory	3.8 kBytes

Table 2: PIC18f24j11 most important parameters [19]

ii. Schematic

The PIC18f24j11 microprocessor has to communicate with both ADS1292R and *Bluetooth* modules. Pins 3 (RA1), 4 (RA2), 11 (RC0), 13 (RC2), 14 (RC3), 15 (SDA1) and 16 (SDO1) are used to interface with the ADS1292R chip. And the Pin 17 (TX1) is used to perform the UART communication. This first prototype does not include flow control in the *Bluetooth* communication, so apart from this pin, there are no other ones needed.

Pins 1 (MCLR), 27 (PGC) and 28 (PGD) are used to program the PIC using PICKIT3 [18].

And finally, an external oscillator of 8 MHz has been connected in order to have an exact frequency operation mode.

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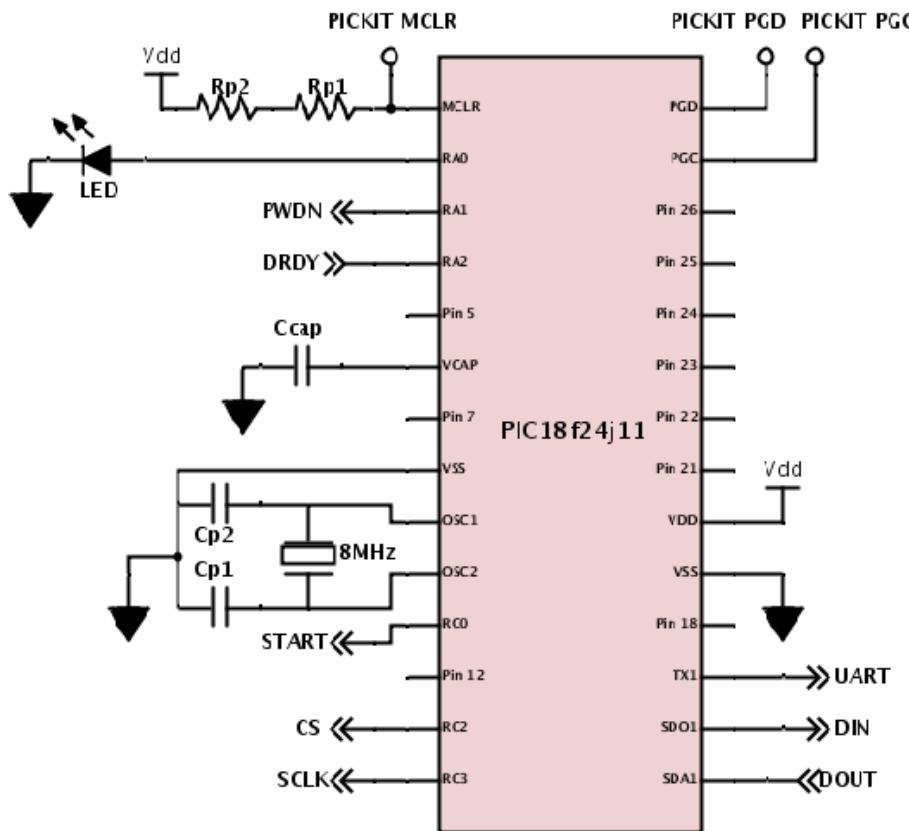
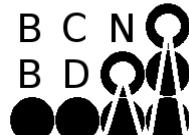


Figure 14: Microchip PIC18f24j11 schematic circuit

iii. Code

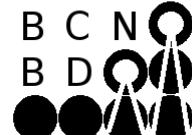
The microprocessor code *main.c* and *main.h* [Annex 1] basically consists of three different parts:

- The configuration of the PIC18f24j11 and the initialization of all needed variables
- The configuration of the ADS1292R using SPI communication
- The reading, processing and resend of all data captured

In the first part (allocated in both *main.c* and *main.h* codes), the configuration of the PIC, the SPI and UART-RS232 communications have to be configured and all needed variables are created. And the frequency is fixed at 8 MHz by an external crystal oscillator.

The SPI communication is used to connect the microprocessor to the ADS1292R. So, the PIC is configured as the master and the frequency clock is divided by 4 (period time large enough for the ADS1292R). The RS232 communication is at ports 17 (TX1, transmitter) and 18 (RX1, receiver). However, the prototype made up in this project does not include flow control. So, the only pin used is TX1. The Baudrate is fixed at 57600 bauds in order to get a sampling frequency of the ECG and respiration signals at least of 500 Hz. Otherwise, at lower frequencies, the ECG signal may have aliasing effects.

The configuration of the ADS1292R is executed in the first part of the code and only once. Future improvements could execute codes that might change this configuration in real time depending on the inputs of the user.



Most of the registers written have the default value except the register two that has the bit PDB_REFBUF activated in order to have the internal reference buffer powered.

The reading, processing and resend code is the one that is indefinitely iterating. Firstly, the three status registers of the ADS1292R, the three bytes of respiration and the three ones of ECG are captured. It is important to notice that this data is in Ca2 format. Consequently, before representing these signals, a conversion from Ca2 to decimal formats has to be performed: this is done in this code. And finally, this data is written in the UART TX port to transmit it to the *Bluetooth* module.

e. *Bluetooth* Module

To perform the *Bluetooth* communication, a commercial module has been selected: CB-OEMSPA311i-04 [19]. It has the following characteristics (Table 3):

Parameter	Value
Internal Antenna	Yes
Serial Port Adapter	Yes
Radiated Power	5mW (7 dBm)
Supply Voltage	3-6 V
Power Average Consumption Transmitting @3.3V @115 kbit/s	72.6 mW
Size	36x16x3 mm
Receive Sensitive Level	-90 dBm
Maxim Receive Input Level	15 dBm
Output Frequency	2.402-2.480 GHz, ISM band
Bluetooth Qualification	01/02/00
Bluetooth Profiles Supported	Generic Access Profile (GAP), Serial Port Profile (SPP), Dial-up Networking Profile (DUN GW, DUN DT)

Table 3: Main characteristics of the CB-OEMSPA311i-04 [20]

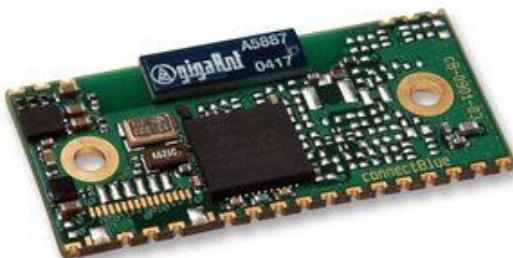
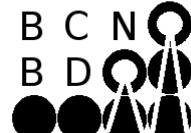


Figure 15: CB-OEMSPA311i-04 [19]

In this project, no flow control has been implemented in the RS-232 or in the *Bluetooth* communication. So, the only Pins connected are the followings:



- Pin 2 (Ground)
- Pin 3 (Vsupply, 3.3 V)
- Pin 11 (Rx, connected to TX Pin of the microprocessor)

f. Labview

The *Labview* code is the element that completes this project. It has to receive the data sent out by the *Bluetooth* module, process it and represent both ECG and respiration signals. All *Labview* code is shown at Annex 2.

First of all, the serial *Bluetooth* communication has to be configured. As stated before, the baud rate is 57600 bauds (in order to not have *aliasing* effects), there is no parity bit and there is one stop bit. Furthermore, there is no flow control. However, in future implementations this should be designed as the product would achieve better performance.

Then, the three bytes of respiration and the three ones of ECG have to be captured. To identify each of them, one synchronization byte is sent: a 0x55. So, firstly, the *Labview* code waits until it detects a 0x55 data byte and then, it starts capturing the three bytes of respiration and ECG (in this order).

Obviously, before plotting the signals, some processing has to be programmed. As the signal is codified by three bytes, they have to be properly concatenated and adapted to represent correctly the magnitude measured. But, this is not all, both signals have to be filtered in order to eliminate those undesired frequencies – high frequencies got from the digital circuit and the 50 Hz interference got from the electrical network. Consequently, some digital filters are included in the program:

- The respiration signal is second order bandpassed filtered between 0.05 and 2 Hz as the signal frequency spectrum is basically in that range. Therefore the 50 Hz interference and high frequencies are eliminated as well as the baseline.
- The electrocardiographic signal is also filtered but in this case between 0.02 and 40 Hz. It is important to notice that despite a second order filter is used at 40 Hz, the 50 Hz still appear at the output. So, a five order bandstop filter is also introduced in series between 45 and 55 Hz in order to completely eliminate this interference.

It is important to realize that despite the sampling frequency is 500 Hz, none of the signals go up from 40 Hz. There are two main reasons:

- In this project, it is only plotted the ECG signal that has a frequency spectrum between 0.02 and 40 Hz but, in fact, it can be until 150 Hz. So, future improvements might use higher frequency bands than the one that is currently used.
- The other reason is that having a higher sampling frequency, it is possible to filter better at higher frequencies.

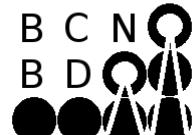
Finally, after having captured, processed, filtered and adapted this input data, both respiration and ECG signals are prepared to be plotted.

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Obviously, this *Labview* code is executed in a PC and not in a smart phone. This means that future improvements might implement a code able to run in a mobile phone.

This final point, when all data is finally captured, opens a huge range of possibilities and properties. A protocol that connects to a hospital when something strange or dangerous is detected on a patient could be implemented. Or maybe, when a car driver's drowsiness is detected, a noisy alarm could be activated.



9. SYSTEM IMPLEMENTATION

a. First Prototype – Perboard

Figure 16 shows the first prototype of the product implemented in perfboard divided in five main parts. Part 1 is the commercial *Bluetooth* module mounted on the evaluation board. Part 2 is a MAX3232 circuit that converts the TTL signal from the microprocessor to RS232. Part 3 is the ADS1292R electronic circuit. Then, part 4 is a TTL to RS232 converter to connect the microprocessor to a computer and, by this method, the code functionality can be easily checked. And finally, part 5 is the PIC18f24j11 electronic circuit.

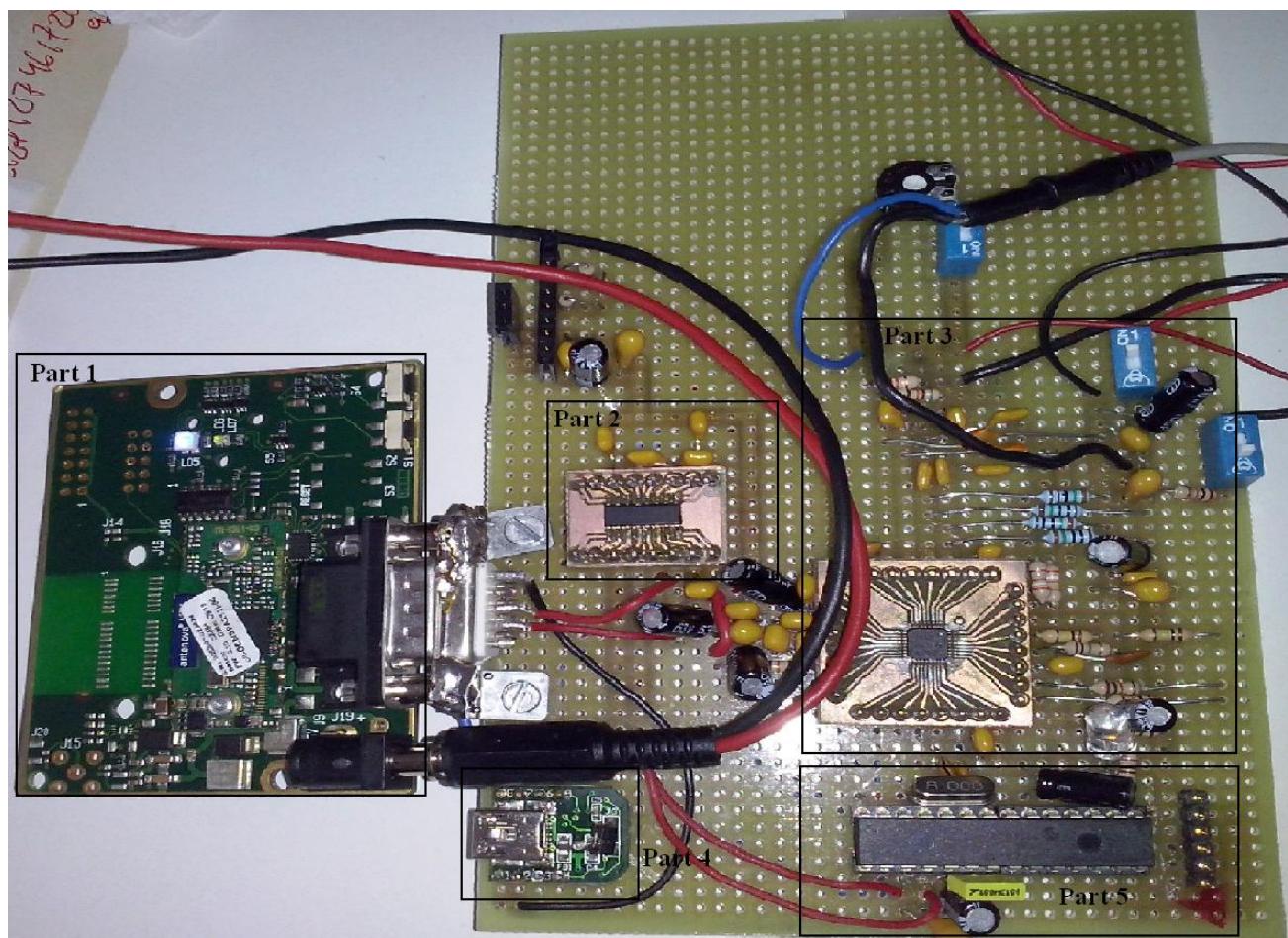
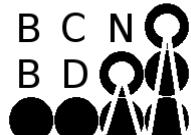


Figure 16: First prototype

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b. Second Prototype - Printed Circuit Board

Figures 17 and 18 show the front and back sides of the PCB prototype. They are designed using Altium with two layers. Digital traces have a width of 0.2 mm (the minimum that the fabrication process allows) and the power and analog lines are as short and thick as possible. Every trace is separated to any other one with a distance at least of 0.2 mm. And its total size is 49 mm long and 57 mm width.

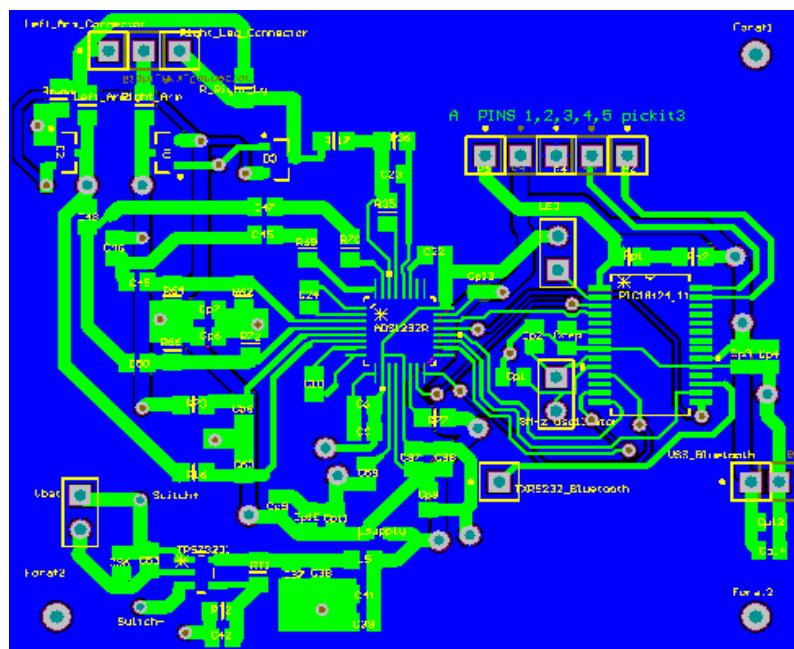


Figure 17: Front PCB side

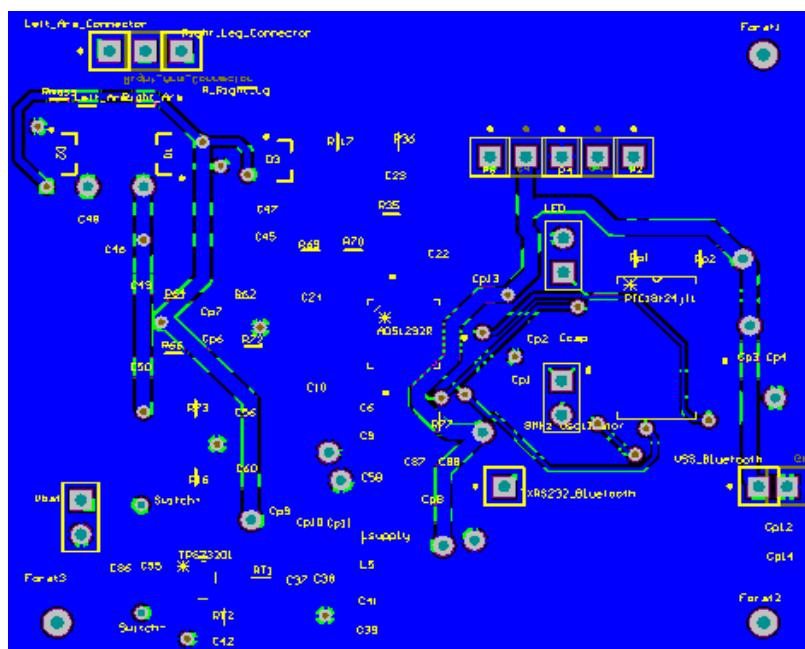
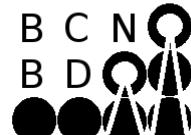


Figure 18: Back PCB side



10. SYSTEM CHARACTERIZATION

a. Patient Simulator

First tests have been done connecting the PS420 Simulator Patient [21] to the first project prototype (perfboard). The results are absolutely satisfactory and both signals have got enough SNR to provide the expected information from both magnitudes. However, in the ECG graphic, it is easily noticed the electrical network frequency interference of 50 Hz. In the second prototype (in PCB) this is improved.

i. Electrocardiography

In Figure 19, the ECG of the simulator is shown. The patient simulator was configured at 80 bpm and 2 mV of amplitude using the Right Arm, Left Arm and Right Leg equivalent inputs. The Y-axis is in mV and the X-axis is in seconds representing two seconds in total.

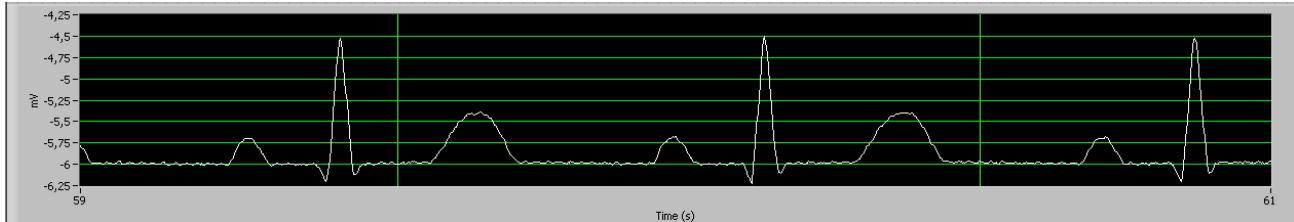


Figure 19: ECG signal got from the PS420 Simulator Patient

ii. Respiration

In Figure 20, the respiration signal of the simulator is shown. The patient simulator was configured at 80 respiratory cycles per minute and only using the Right Arm and Left Arm equivalent inputs. The Y-axis is obviously in ohms (it is being used a BEI) and the X-axis is in seconds representing two seconds in total.

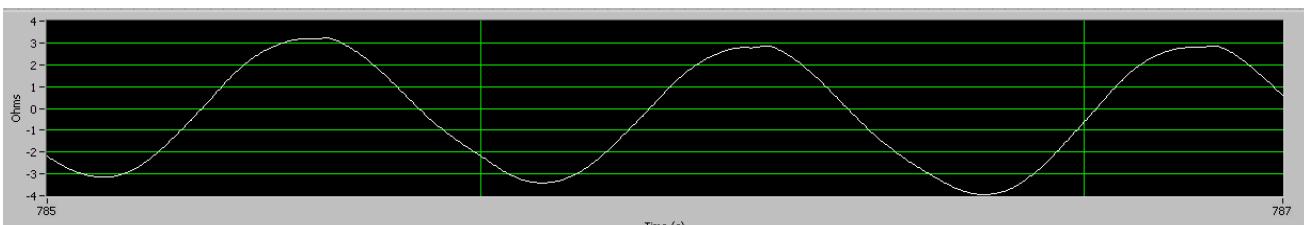
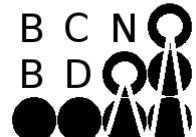


Figure 20: Respiratory signal got from the PS420 Simulator Patient



b. Real Human Volunteer

When all tests using the patient simulator were performed, the device was tested with real humans. The following figures show the electrocardiographic and respiration signals during two seconds of a volunteer that had two electrodes connected at the two sides of the thorax.

i. Electrocardiography

The electrocardiographic signal represented below has the expected wave with some little differences as the position of the electrodes was not the one commonly used to capture the ECG.

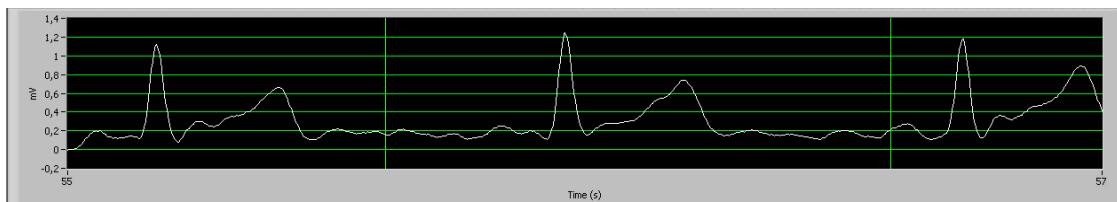


Figure 21: ECG of a real human volunteer

ii. Respiration

In Figure 22, two inspirations and exhalations are absolutely clear.

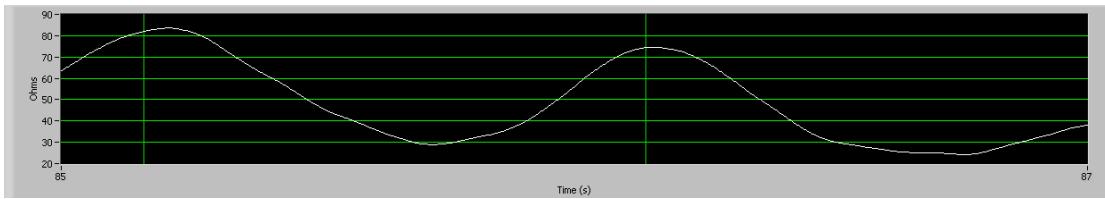
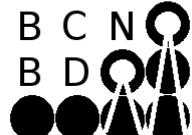


Figure 22: Respiration of a real human volunteer



11. BUDGET

a. Product Components Cost

The total components cost of a single product is deeply detailed in Annex 3.

The cost depends on the quantity of products that are going to be manufactured:

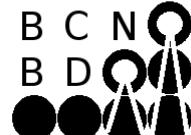
- Manufacturing one: 39,26 €/product
- Manufacturing one hundred: 27,45 €/product
- Manufacturing ten thousand: 23,83 €/product

b. Design and Prototyping Costs

Concept	Price (€)
Engineering cost (600 hours · 20 €* · 1 person)	12000
Evaluation board ADS1292RECG-FE	74
Installations amortization (PC, Lab facilities...)**	250
Design components	20
First prototype components and perfborad cost	44,36
Second prototype components and PCB cost	68,36
Total	12456,72

*Junior engineer contract

**200 € for one PC amortization during one year plus 50 € in other lab facilities amortization



12. CONCLUSIONS

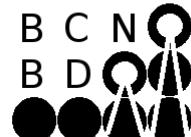
In this project, the electrocardiographic and respiratory signal characteristics have been deeply analyzed. The extracted results have provided essential information about what exactly the final product should be. This information has also shown critical aspects used in the design, implementation and testing of the prototype.

When the general structure was totally clear, it was the moment to specifically design each subcircuit considering all the connections of the final architecture. Every module has been independently conceived and designed. Even the ADS1292R, which is the most important component of the general structure, was first tested before being implemented in the final whole circuit to be absolutely sure that it perfectly suited the purpose of the project.

Step by step progress has been achieved. Starting from the integration of ADS1292R and microprocessor, the remaining modules have been added. The independent test of each subcircuit has become a difficult goal that in a way has delayed the project. Isolate every part of the circuit and depurate its functionality has not been easy as every module needed the performance of the others ones.

There are several applications that could use this product for different purposes: drowsiness detection, sudden cardiac death sensing, airway obstruction alarm or monitoring other pathologies. Any of these services could offer important benefits to the society as saving lifes or improving the wellness of the population. As stated in the introduction, multiple research groups are investigating in the same concept. There are different technologies but they all go after the same objective: monitor the respiration or the electrocardiogram.

The objective of this project was to develop a product that measures the electrocardiographic and respiration signals and represent them to the user. This has been absolutely achieved as test results show. However, there are still many possible improvements to design and implement in order to manufacture a better final user product. Moreover, deeper analysis and tests have to be done. But this project provides a fundamental start point for future improved applications.



13. REFERENCES

- [1]: Drowsiness Detection by Thoracic Effort Signal Analysis with Professional Drivers in Real Environments. N. Rodríguez, M. García, J. Ramos, M. Fernández 33rd Annual International Conference of the IEEE EMBS
- [2]: Contactless electrical bioimpedance system for monitoring ventilation. A biodevice for vehicle enviroment. R. Macías, M. García, J. Ramos, R. Bragós, M. Fernández SciTePress
- [3]: HRV based Health&Sport Markers Using Video from the Face. L. Capdevila, J. Moreno, J. Movellan, E. Parrado, J. Ramos 34th Annual International Conference of the IEEE EMBS
- [4]: www.facelake.com/contactus.html Access date: 9/11/2013
- [5]: www.alivercor.com Access date: 9/11/2013
- [6]: http://www.analog.com/static/imported-files/data_sheets/AD5933.pdf Access date: 12/11/2013
- [7]: Current Source for Multifrequency Broadband Electrical Bioimpedance Spectroscopy Systems. A Novel Approach. F. Seoane, R. Bragós, K. Lindecrantz 28th IEEE EMBS Annual International Conference
- [8]: An analog front-end enables electrical impedance spectroscopy system on-chip for biomedical applications. F. Seoane, J. Ferreira, J. Sánchez, R. Bragós Physiological measurement
- [9]: Biomedical Instrumentation Systems. S. Chatterjee, A. Miller
- [10]: <http://en.wikipedia.org/wiki/Electrocardiography> Access date: 12/11/2013
- [11]: http://www.cardiologyjournal.org/en/darmowy_pdf.phtml?indeks=86&indeks_art=1123 Access date: 12/11/2013
- [12]: http://general.utpb.edu/fac/eldridge_j/Einthoven%20Tri.htm Access date: 12/11/2013
- [13]: Medical Devices and Systems – The Biomedical Engineering Handbook Edited by Joseph D. Bronzino
- [14]: <http://www.clevelandclinicmeded.com/medicalpubs/diseasemanagement/pulmonary/pulmonary-function-testing/> Access date: 12/11/2013
- [15]: <http://www.ti.com/lit/ds/symlink/ads1291.pdf> Access date: 12/11/2013
- [16]: <http://www.ti.com/lit/ug/slau384a/slau384a.pdf> Access date: 12/11/2013
- [17]: <http://www.ti.com/lit/ds/symlink/msp430f5529.pdf> Access date: 12/11/2013
- [18]: http://ww1.microchip.com/downloads/en/DeviceDoc/PICkit_3_User_Guide_51795A.pdf Access date: 13/11/2013
- [19]: <http://ww1.microchip.com/downloads/en/devicedoc/39932c.pdf> Access date: 13/11/2013
- [20]: http://www.connectblue.com/fileadmin/Connectblue/Web2006/Products/Bluetooth/Gen3/M0M1_M2/OEM/Common/Documents/E_M_Datasheet_OEMSPA_311_331.pdf Access date: 13/11/2013
- [21]: http://assets.fluke.com/manuals/ps420_umeng0100.pdf Access date: 16/11/2013