

Final Degree Thesis

***AD5933-Based System for
Transthoracic Impedance
Measurements***

By

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AD5933-Based System for Transthoracic Impedance Measurements

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Master Degree Thesis

Subject Category: Medical Technology, Signal Processing

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Date: 31th August 2012

Keywords: Electrical Bioimpedance, AD5933, VB.NET, Matlab, C.

ABSTRACT

Electrical Bioimpedance (EBI) through transthoracic measurements offers a non-invasive manner to measure hemodynamic values. In the last decades, advances in the field of electrodes, electronic hardware and the development of textrodes have paved the ways for smaller, easier to use equipment for EBI measurements.

In this thesis a AD5933-based board originally designed for full-body edema monitoring has been adapted to perform continuous Transthoracic EBI measurements. For this objective 2 pieces of software have been developed: a firmware in C to control the board and a user interface in Visual Basic 2008 for Windows to easily monitor and record captured data.

This thesis also presents the results of the validation tests carried out onto 2 test subjects. These results have been exported and analysed with MATLAB and show the feasibility of using this system for extended monitoring but no conclusive data can extracted without a further analysis.

ACKNOWLEDGEMENTS

Quiero agradecer a mis padres y mis amigos con un testimonio de eterno agradecimiento por el apoyo moral que desde siempre me brindaron y con el cual he logrado terminar mi carrera académica, que es para mí la mejor de las herencias.

Con cariño y agradecimiento.

Víctor Serres Altés.

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LIST OF ACRONYMS

EBI	-	Electrical Bioimpedance
V/I	-	Voltage / Current
ICF	-	Intracellular Fluid
ICG	-	Impedance Cardiography
ECF	-	Extracellular Fluid
EEG	-	Electroencephalogram
EKG	-	Electrocardiogram
TEB	-	Transthoracic electrical bioimpedance
UI	-	User Interface

1.1 Introduction

Bioimpedance is the electrical impedance of a biological tissue. Bioimpedance characteristics and value depends on the type, structure, composition, behavior and any changes over time that might occur to the tissue.

Impedance cardiography is the study of the bioimpedance of the human body but limiting the measurement area to the chest cavity. The changes of the bioimpedance signal can be monitored to estimate cardiac parameters and assess the cardiac cycle. The monitoring can be performed using non-invasive measuring methods with skin electrodes.

1.2 Motivation

Patient comfort, mobility, monitoring and cost-effectiveness are all desirable characteristics of a measurement system. A portable system using AD5933 has already designed by Javier Ferreira for performing bioimpedance spectroscopy measurements aiming for body composition analysis. The intent of this thesis is the adaptation of the existing board to perform Impedance Cardiography ambulatory monitoring. Such system could enable ICG monitoring in e-health applications.

The main motivation of this thesis is to save development costs when designing the new system by reusing the existing system and developing a new firmware to perform continuous monitoring and recording of the ICG signal.

1.3 Goal

The primary goal is the development of a firmware for the bioimpedance measurement system and a Windows application capable of performing continuous monitoring and recording of a subject ICG signal. The windows application must be able to store all the measured data, display it onscreen and export it to a file. The data collected must be formatted so it can be easily imported by third party software like MATLAB.

A secondary goal is to optimize the firmware so it can work with the maximum measurement sampling rate of the AD5933 to allow the time resolution required to perform ICG signal analysis.

1.4 Work done

A firmware and a windows application have been developed as the main work of this thesis. The firmware, written in C language, allows total control of the board using a command line interface accessible via Bluetooth or USB serial ports. The application, developed with Visual Basic 2008, is designed to be capable of adjusting the measurement system prior measurements, initiating and stopping measurement sessions. The application is also capable of displaying the data collected using a user friendly interface with zooming and

panning capabilities. The data collected can be exported to a text file for further processing with other tools.

The maximum sample frequency to perform a single impedance measurement has been study to ensure that the AD5933-based bioimpedance system can be used to obtain an accurate ICG measurement.

1.5 Structure of the Thesis Report

The whole thesis report is organized in six chapters with references. Chapter 1 is the introduction part of the thesis. Chapter 2 gives a brief theoretical background of the subject of bioimpedance, the studies done in the subject and how ICG can help in determining heart cycle parameters. Chapter 3 deals in detail with board and its components while chapter 4 describes thoroughly the work done and its functionality with the help of several screenshots. The chapter 5 discusses briefly the results obtained testing the new board firmware and the chapter 6 presents the conclusion and deals with possible future works.

1.6 Out of Scope

This thesis deals exclusively with the implementation of the code for adapting the board to the new application and presents the results in an illustrative manner to show that the work done functions properly. Although the electronics from acquiring the EKG were not available in the original system, the design and implementation of the EKG amplifier is not dealt with in here, such task has been the implemented in a different thesis work.

In the field of medicine, the amount of information that can be obtained from a patient helps in the process of diagnosing the patient illness. In cases of emergency, when the patient is critically injured or presents an unknown disease that threatens his or her life, doctors rely on the vital signs to monitor the status of the patient and to diagnose the illness. Nowadays the four vital signs that are standard in every medical procedure are: the electrocardiogram or EKG which shows, among other factors, the cardiac pulse; the blood pressure, the respiration rate and the temperature.

As the amount of information available for the doctors eases the diagnoses of patients, as such some authors have studied and verified the utility of employing pulse oximetry as the fifth vital sign [4, 5]. Moreover there are authors that defend impedance cardiography (ICG) as a suitable candidate to be used as a sixth vital sign[6].

2.1 Bioelectricity and Bioimpedance

The concepts of bioelectricity and bioimpedance don't differ much from the more classical electricity and impedance concepts.

Bioelectricity concept is more general and encompasses all electric phenomena in a living body both in animals and plants. The effects of these phenomena vary according to voltage, current, frequency and/or type of the tissue. One classification of the electrical phenomena relevant to this thesis is the distinction between the ones caused by exogenic, or externally applied, electricity like in the case of bioimpedance measurement and the ones caused by endogenic, or internally generated, electricity like muscle excitation, nervous system and brain activity or cardiac impulses. Prime examples of the latter set of phenomena would be EKGs and EEGs.

The bioimpedance of biological tissue express the capacity of a tissue to *impede* (oppose) the flow an electric current flowing through it. Bioimpedance values depend on a wide range of factors including if the tissue is alive or dead and the presence and quantity of fluids. Its value typically presents two parts: the resistance and reactance. The former value seems to depend on the extra- and intracellular fluid while the latter is an effect of the cell membranes [7]; in fact, in general terms is good to presume that most of the tissues are anisotropic instead of isotropic [8].

2.1.1 A Brief History of the Fields of Bioelectricity and Bioimpedance.

The idea of using electricity to assess the status of a living being is not new; Carlo Matteucci (1811-68) was the first to measure a bioelectric current measuring the effects of electric current in the muscle of a frog in 1838. Some years after in 1887, the British

physiologist Augustus Waller (1856-1922) was the first to measure the electric activity of the heart and made the first register EKG. In 1924 the German Hans Berger (1873-1941) performed the first electroencephalography of a human brain.

In 1872, Thomas Green successfully reanimated patients who suffered respiratory and cardiac arrest, by using a battery made up of 200 cells, which could generate up to 300V. In 1899, the first cardiac defibrillation was reported by Prevost and Battelli who discovered that while low voltage shocks induced ventricular fibrillation high voltage shocks did the opposite effect[1].

The first author to start researching the fields of bioimpedance before it was known as such was Hermann von Helmholtz with the conduction velocity of a nerve cell axon around 1850. In the 1870's Hermann Müller found the capacitive properties of tissue and anisotropy of muscle conductance. James Clerk Maxwell announced the joint equations set the foundation of electrodynamics (1861) and developed the mathematical model for cell suspensions and tissues that it's still in use today.

It was in 1940 when Kenneth S. Cole presented the well-known Cole equation and in 1941 when both brothers, Kenneth and Robert, presented the also known Cole-Cole equation which will be explained more closely later, both equations are shown in 2.1 and 2.2.

$$Z = R_{\infty} + \frac{R_0 + R_{\infty}}{1 + (j\omega\tau)^{\alpha}} \quad (2.1)$$

$$\varepsilon^* + \varepsilon_{\infty} = \frac{\varepsilon_0 - \varepsilon_{\infty}}{1 + (j\omega\tau_0)^{1-\alpha}} \quad (2.2)$$

It was not until 1962 when Herman Paul Schwan released his pioneering research with four electrode techniques and gigahertz measurements [8].

2.1.2 Electrical Properties of the Biological Tissues

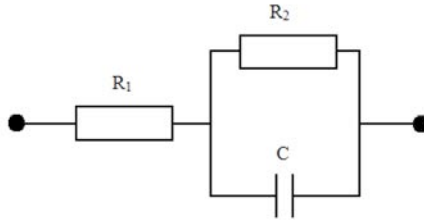
Bioimpedance or resistance to the flow of the electric current through a biological tissue is the most basic of the electrical properties of a material and depends on the presence of different constituents like for example ions, cells or fluids among others. The resistivity is determined by type of tissue, for example blood plasma has the lowest resistivity being 63 Ωcm and the highest one is observed for fat 1,275 Ωcm , see Table 1.

Table 1- Typical resistivity values for different tissues, source [2]

Type of the tissue	Resistivity (Ωcm)
Blood plasma	63
Blood (for hematocrit Ht = 47%)	150
Skeletal muscle (longitudinal)	300
Skeletal muscle (transverse)	700
Heart muscle (dog)	750
Lungs (dog)	1,200
Fat	2,180
Saline 0.9%	57

Tissue as a Conductor

Tissues can be treated as a finite, inhomogeneous volume conductor and we can also assume that they are a source less medium. The tissue impedance can be represented by a three-component electrical model with a single time constant, see Figure 1.

**Figure 1-** Biological Tissue electrical model

For this model represented in Figure 1, the Impedance value can be represented using the Cole Function as shown in Equation 2.3.

$$Z(f) = R_{\infty} + \frac{R_0 - R_{\infty}}{1 + (j\omega\tau)^{\alpha}} \quad (2.3)$$

In Equation 2.3, $Z(f)$ is the impedance as a function of frequency, R_0 resistance at frequency zero, R_{∞} resistance at infinite frequency, τ time constant RC and the parameter α that for Electrical Components has a value 1 and for tissues a value between 1 and 0.

In impedance cardiography, the measurement process involves using a single excitation frequency, normally 50 kHz, to perform synchronous impedance measurement to obtain the impedance magnitude value over a period of time. Because of the use of the constant frequency, the Cole models are not used in impedance cardiography.

2.2 Bioimpedance Measurement Equipment

The required equipment to perform bioimpedance measurements typically consists of a module capable of injecting a controlled signal and to sense the response signal to calculate the estimated impedance.

The interface between the measurement device and the body under study is typically built up by a set of skin electrodes. The electrodes allow the interface between the tissue and electrical components in the measuring device, allowing ionic polarizations to be converted into electron displacements. Conventional Electrodes have the enormous advantage to being a non-invasive method of measurement, which allows the research and development of all kinds of applications including portable applications usable by the patient at home and 24h monitoring systems.

The Sine Correlation is one of the most common techniques used in the field of EBI and also by most of the commercial EBI spectrometry devices. The method involves injecting a signal into the tissue, voltage or current, and then sensing the response that crosses the tissue and processing it. The signal injected to the tissue is a current sine signal with a fixed frequency, and the measured voltage is multiplied by the in-phase signal and the in-quadrature signal in order to estimate the Real and Imaginary components from the Measured Impedance, the sine correlation system blocks are shown in Figure 3.

Current is injected through the injecting electrodes to the body of the patient and then the voltage is sensed between the sensing electrodes. The mathematical expression of the voltage sensed is shown in the equation 2.4.

$$V = Z * I \sin(\omega t) = R * I \sin(\omega t) + X * I \cos(\omega t) \quad (2.4)$$

Equation 2.5 shows the process to obtain the real part of the impedance measurement. It involves multiplying the sensed signal by a sinus of the same frequency to obtain the real part of the impedance as a baseline signal and modulate the imaginary part of the impedance to twice the frequency of the input signal. The last step is application of low pass filter to obtain the resistance.

$$\begin{aligned} (I R \sin(\omega t) + I X \cos(\omega t)) \sin(\omega t) &= I R \sin^2(\omega t) + I X \sin(\omega t) \cos(\omega t) \\ &= \left[\sin^2(\omega t) = \frac{1 - \cos(2\omega t)}{2} \right] = \left[\sin(\omega t) \cos(\omega t) = \frac{\sin(2\omega t)}{2} \right] \\ &= \frac{I R}{2} - \frac{I R}{2} \cos(2\omega t) + \frac{I X}{2} \sin(2\omega t) \end{aligned} \quad (2.5)$$

To obtain the imaginary part of the impedance the same process is followed like it is shown in equation 2.6. Instead of multiplying the sensed signal by the input signal directly the last is phased $\frac{\pi}{2}$ [3].

$$\begin{aligned}
(I R \sin(\omega t) + I X \cos(\omega t)) \cos(\omega t) &= I R \sin(\omega t) \cos(\omega t) + I X \cos^2(\omega t) \\
&= \left[\sin(\omega t) \cos(\omega t) = \frac{\sin(2\omega t)}{2} \right] = \left[\cos^2(\omega t) = \frac{1 + \cos(2\omega t)}{2} \right] \\
&= \frac{I R}{2} \sin(2\omega t) + \frac{I X}{2} + \frac{I X}{2} \cos(2\omega t)
\end{aligned} \tag{2.6}$$

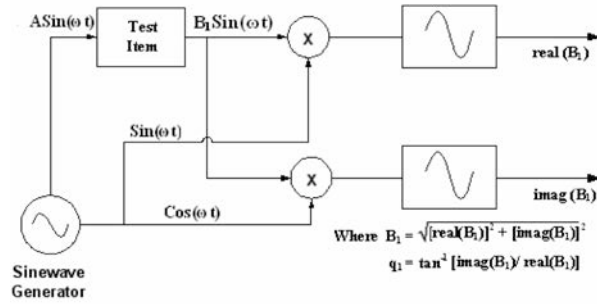


Figure 2. Measurement setup for four-electrode technique with Sine-Correlation analysis.
Source [3]

2.2.1 Bipolar and Tetrapolar Methods

There are mainly two electrode configurations to perform impedance measurements: Bipolar Configuration and Tetrapolar Configuration.

In the Bipolar Configuration two electrodes are used and both play the role of injecting the current and sensing the voltage, see Figure 4. This configuration is problematic because the impedance measurement contains two times the electrode polarization impedance plus the impedance under measurement. In practice, it is difficult to separate them and obtain a clean measurement, but for some applications where an impedance trend is needed this configuration is used.

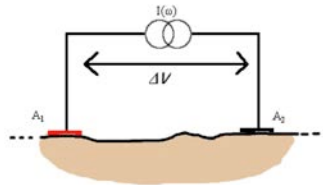


Figure 3- Bipolar configuration

The second method involves the use of four electrodes (tetrapolar), which play different roles each. Two electrodes apply the electric potential and the other two receive the impedance signal, see Figure 5.

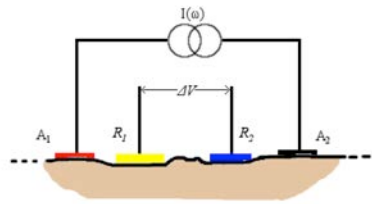


Figure 4- Tetrapolar Configuration

The injecting current is applied via the A_1 - A_2 electrodes while the detected current is measured through the R_1 - R_2 electrodes. This method minimizes the load of the skin-electrode impedance, because under ideal circumstances the input current in the voltage sensing is zero, therefore the measure voltage it is equal to the tissue voltage and the voltage electrodes are zero.

2.2.2 Electrical Bioimpedance Applications

The major advantage of the electrical bioimpedance measurement is the non-invasive measuring method through the use of skin electrodes. A non-invasive measurement translates into a minor or minimal requirement for trained medical personnel to operate the devices and the patients experience far less annoyance during the measurement procedure.

This lack of need for medical personnel to oversee the operation of bioimpedance measurement systems opens a world of opportunities for small and portable equipment that can be operated by users themselves at their homes.

All these advantages have encouraged the development of multiple applications like: Tissue characterization: Measuring the impedance of different patches of skin a range of assessments can be performed like skin hydration levels, status of skin damage both by irritation or allergic sources, hyperhidrosis and skin cancer.

Total body composition: Through measuring the full body impedance and using a wide range of frequencies is possible to calculate the amount of fat mass present in the body and the relation between fat mass and muscular mass.

In addition to total body composition there is other applications like Impedance Plethysmography, which involves measuring the impedance offered by the whole body or a part of it continuously to measure the changes of tissue volume. The measurement of impedance in the chest or thorax region is called *Impedance Cardiography*. One direct application of Impedance Plethysmography is the detection and assessment of tissue edema [9].

2.3 Impedance cardiography

Bioimpedance Plethysmography is a method of studying tissue volume changes in a living body by measuring the changes of electrical impedance at the body surface. Impedance Cardiography (ICG) is the Impedance Plethysmography measurement limited to the chest/thorax region.

Electrocardiography or EKG is the process of recording all the electric excitation phenomena of the heart. The EKG records with a good level of detail the electrical excitation in different parts of the heart muscle: which include the sinus and AV nodes, the atrial and ventricular muscles and nerve fibers in the ventricular muscle; it is crucial in the diagnosing of several heart diseases. As will be seen later on, the EKG signal shows a good relation with the ICG signal as some of the waves of the EKG are correlated with waves in the ICG measurement.

As with EKG, ICG measurements have the potential of being done in a non-invasive way. Several authors have tried to correlate invasive methods like thermo-dilution with ICG with skin electrodes with different degrees of success [10, 11]. Most of the authors agree that ICG requires more methodological investigation but are good enough to estimate changes of the Stroke Volume.

2.3.1 Impedance Cardiography Signal

The impedance cardiography signal is obtained by injecting a current and then sensing the voltage in the thorax region to estimate the impedance using the Ohms law equation (2.4).

$$Impedance = \frac{Voltage}{Current} \quad (2.4)$$

The typically used current to measure the impedance is a sinusoidal signal with a constant frequency between 20 and 150 kHz with a constant amplitude of 0.5-5 mA. The impedance usually consists of a Real component, with values between 15Ω to 30 Ω, and an often negligible and not used imaginary component. The changes of impedance due to the changes of fluid in the chest area are typically between 100-400mΩ which results in a change of 0.1-0.4 mV in the signal received when applying 1mA to the patient.

Blood is characterized by having a very high conductivity being its resistivity of 130-160 Ωcm two times lower than muscle tissue. Moreover, blood resistivity is many times higher than the resistivity of other tissues. The blood resistivity plays a major role in the measurement of cardiac performance using impedance methods. Initially, there has been a general assumption that the blood resistivity has isotropic properties in static conditions but later measurements have led to suggestions that flowing blood is characterized by a relative resistivity decrease of a maximum 10%. These facts are what draw the conclusion that the impedance measured between the receiving electrodes is the impedance caused by the blood volume contained within the segment of chest between electrodes.

Several attempts have been performed to create a model that could realistically represent the impedance fluctuations and with it investigate the origin of the impedance signal. The first

attempt was simplify the chest region and represent it as a cylinder partially filled with blood where blood changes would be represented as variations of diameter, for instance the blood ejection from the heart would be represented as an increase of diameter, see Figure 7.

The validity of the one-cylinder model was questioned after numerous inaccurate measurements. Further studies developed more complex models like a multi-cylinder or even more realistic models featuring representations of the main blood vessels.

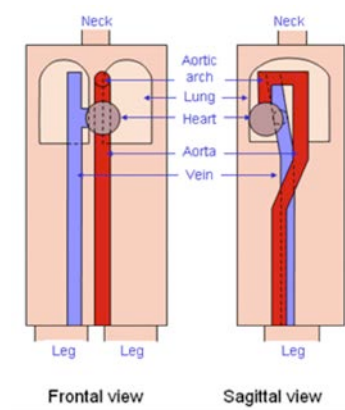


Figure 7 Thorax model by Sakamoto et al. (1979). Source[1]

With the help of the aforementioned theoretical models studies have determined that the following mechanical phenomena of the cardiac cycle have an impact in the impedance cardiography signal:

1. The aorta and neck artery's extension caused by arterial pulse pressure
2. The extension of the pulmonary vessels
3. Changes in the volume of the heart and the volume of blood filling it
4. Changes in the blood filling the pulmonary vessels resulting in an increase of the conductivity of the lungs
5. Changes in the blood resistivity in large vessels caused by reorientation of the blood cells due to the blood flow
6. Changes in skeletal muscle resistivity caused by the pulse blood flow

Several studies have been conducted with the aim of determining the origin of the impedance signal. Penney [12] summarized the contribution of anatomical structures into the impedance cardiography signal, see Table 2.

Table 2- Contribution of organs on the impedance cardiography signal

Contributing organ	Contribution (%)
Pulmonary artery and lungs	63
Aorta and thoracic musculature	150
Right ventricle	300
Left ventricle	700
Pulmonary vein and left atrium	750
Vena cava and right atrium	1200

Mohapatra[13] after analyzing several hypotheses regarding the impedance cardiography signals concluded that the impedance changes were caused only by cardiac hemodynamics, i.e. changes in blood volume and blood velocity. Moreover, the variable speed of ejection has a major effect on the systolic part of ΔZ while the changes in volume affect the diastolic part of the impedance curve[13].

With the papers of Mohapatra the following conclusions can be extracted[6]:

1. The ICG signal is complex and its components are not synchronized in phase and direction
2. The proportions between its components are not well defined
3. The ICG signal is dependent on the changes in the diameter of the aorta, neck arteries and pulmonary vessels and in the resistivity of flowing blood caused by blood cell reorientation (only in the large vessels)
4. The ICG signal appears not to be dependent on heart volume and the amount of blood in it (isolating properties of the heart muscle in relation to blood resistivity)
5. For typically positioned electrodes the ICG signal is not dependent on pulmonary artery extension

These conclusions, while useful, are not enough to discover the origin of the impedance signal is not still clearly known. All the measurements previously described show a high correlation with the events are related to but there is still controversy between most authors and further research has still to be conducted.

2.3.2 Electrode Placement

The electrode is the part of the measurement system where electrical currents shift from electronic conduction to ionic conduction, this is, the currents shift from crossing a metallic conductor to an electrolyte gel or tissue liquid.

Initially Kubicek [14] suggested the use of band electrodes, that offered lower impedance having a bigger effective area than the other electrode types[8]. Unfortunately band electrodes were also cumbersome and difficult to handle so there was a change to use spot electrodes, which come in a variety of formats, such as disposable single use electrodes. In the last few

years, advances in textile technology have increased the viability of textile electrodes for impedance cardiography [15]

Following what has been discussed earlier, the most accurate method to make measurements is the tetrapolar method involving 4 electrodes. The most common electrode placement is the one described by Kubicek in 1966 using, see Figure 8, band electrodes and putting them two at the neck one at the xiphoid point of the sternum and another at the base of the ribcage.

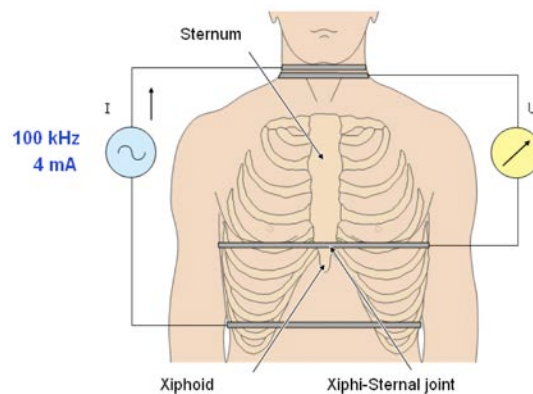


Figure 8 Kubicek electrode placement. Source[1]

In Figure 8, the top and bottom bands inject the current while the two middle bands recover the current that is used to perform the impedance calculation according to the Ohm law.

Although being less noisy having lower impedance [8] than other kind of electrodes electrode bands were known for having handling problems [6] and were slowly being replaced by spot electrodes of different types.

2.3.3 Impedance Cardiography Signal Analysis

Typically the impedance signal follows a known pattern showing characteristic points that help in analyzing the signal and interpreting the results.

As an example, Figure 9 presents 4 typical graphs of the measurements of impedance cardiography (Z), its derivative (dZ/dt), an electrocardiogram signal (EKG) and a phonocardiogram measurement (PCG). The PCG is here included for the sake of completeness and to provide a reference with a well understood technique. The PCG is not currently used in ICG research.

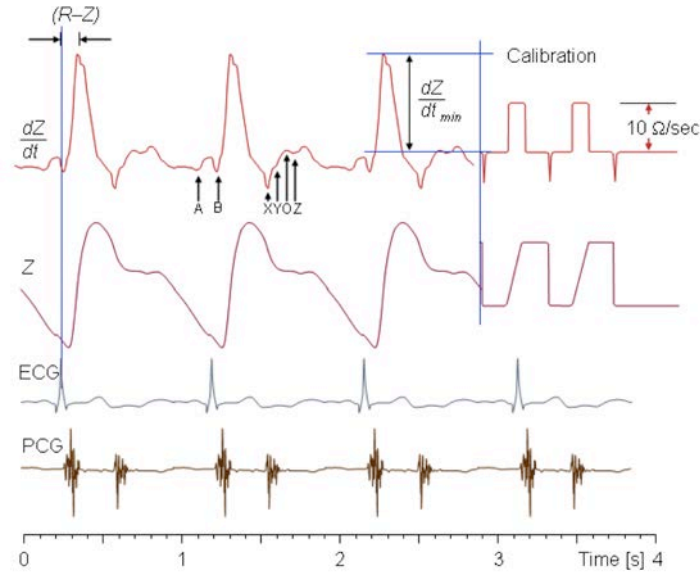


Figure 9 Representation of different kind of measurements of the same heart. Source [1]

The derivative of the impedance signal is the signal used for calculating the hemodynamic parameters like the cardiac output. In the derivative signal there are some points of the heart cycle that can be identified and are marked with the letters A,B,X,Y,O and Z.

Characteristic points in the Impedance Cardiography Graph

Lababidi et al.[16] used phonocardiogram measurements coupled with ICG measurements to associate some of the notches in the dZ/dt curve with different mechanical events of the cardiac muscle.

Table 3- Characteristic points in the Impedance Cardiography Signal

A (atrial wave)	This wave is associated with the atrial contraction and its amplitude is correlated with the ejection fraction of the left atrium. This wave is not observed in case atrial fibrillation.
B	This wave is associated with the opening of the aortic valve. It occurs on the ascending part of the ICG curve before $(dZ/dt)_{\max}$. The precise identification of a B point is important for accurate calculation of systolic time intervals, stroke volume, and cardiac output
E $(dZ/dt)_{\max}$	It is associated with the maximal velocity of the ejection measured by using ultrasound methods. There is some confusion regarding the denomination of this point.
X	Corresponds to the closure of the aortic valve (and second heart sound) and is shown as a minimum value of the dZ/dt signal after the E value.
Y	Corresponds to the closure of the pulmonic valve
O	Associated to the change of volume during diastolic phase of the cardiac cycle and opening snap of the mitral valve.
Z	This wave is associated with the heart sound.

Table 3 associates each characteristic point to its best known originating heart cycle event. The points are best viewed in the ICG signal collected from a healthy patient while laying either in supine or prone position.

Table 4- Characteristic periods in the Impedance Cardiography Signal

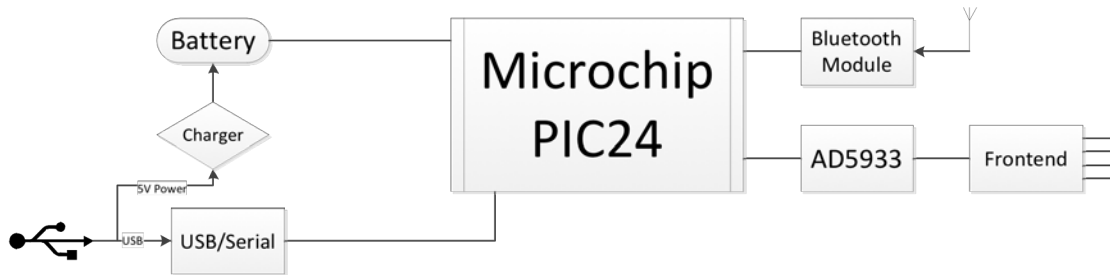
Q-B	Time between the onset of the electrical contraction of the chambers (represented as Q point in the EKG) and the opening of aortic valve. This period is also described as pre-ejection period (PEP)
B-X	Time between the opening and closing of the aortic valve described as left ventricular ejection time (or simply ejection time ET)

The two periods shown in the Table 4, PEP and LVET, are the hemodynamic parameters estimated from ICG signal most frequently used since they have direct association with the cardiac systole.

This chapter reviews and describes the methods followed and materials used for the completion of the thesis work. This chapter will focus primarily on the impedance measuring unit used in the thesis and also the measurement methods used to validate the impedance synchronous measurements for Impedance Cardiography.

3.1 Measuring Unit Hardware

The functional blocks of the impedance measuring unit is shown in the Figure 1. The impedance measuring unit consists of a PIC24 architecture microcontroller from Microchip Inc., an USB port, a Bluetooth transceiver chip, the impedance measuring unit is based in the AD5933 impedance converter IC and the Analog Frontend that enables a 4-wire Bioimpedance Measurements. The whole system is powered by a Lithium Polymer (LiPo) battery that could be charged by a standard USB host controller.



3.1.1 Impedance Network Analyzer AD5933

The AD5933 is a 12-bit high precision Impedance Network Converter from Analog Devices[17] that includes both an onboard frequency generator and a 12-bit 1 MSPS Analog-to-Digital converter to perform the impedance estimation. The AD5933 features an I²C interface to allow the chip configuration and the communications with an I²C Master Controller.

The AD5933 features 1 channel to perform the impedance measurements, one pin is used to inject the voltage sine wave signal and other pin to read the input current in order to estimate the impedance magnitude and Phase values. The output voltage signal is sinusoidal with a frequency range from 1 kHz to 100 kHz. The range of impedances the chip can measure is from 1k Ω to 1M Ω . The impedance conversion time is approximately 1ms when a 16 MHz oscillator is connected to the AD5933 chip.

In order to obtain the magnitude and phase values of the connected impedance, first the output signal is generated and after a programmed time the chip performs a 1024 point DFT on the digitalized signal received from the impedance, then the result is placed in two separate registers. To obtain the real and imaginary values of the impedance, the obtained results need to be multiplied by a calibration factor that it has been previously obtained.

Measurement and Calibration process

The measurement process for the AD5933 is very streamlined and has almost no complexity. In the case of a single frequency measurement the procedure is even easier as the AD5933 only requires the starting frequency value and then it is ready to be issued the command to start measuring.

The AD5933 requires to be adjusted for the proper accuracy before any proper use. The value measured by the AD5933 is not the value of the impedance and must be corrected before use. Adjusting the accuracy of the measuring system is done by calculating the so-called Gain Factor, which is simply achieved by measuring a known impedance to obtain the magnitude value and by using the magnitude value of the connected impedance the Gain Factor can be obtained, see equation 3.2. The gain factor calculated will be multiplied to the magnitude value measured on every measurement to obtain the magnitude value of the connected impedance.

$$\text{Gain Factor} = |\text{Known Impedance}|/|\text{Measured } Z| \quad (3.2)$$

3.1.2 Analog Front-end for Bioimpedance Measurements

The AD5933 IC is an excellent solution that incorporates all the necessary devices to perform an impedance measurement in a Single Chip Solution, but also suffers of several drawbacks that make it not suitable for the measurements of Bioimpedance. Previous studies [9] have implemented an Analog Front End (AFE) that enables the AD5933 for the performance of EBI measurements.

The Analog Front End [9] enables the conversion of 2-wire to 4-wire measurements. The constant output voltage is converted to DC-free and controlled constant injecting current that has been designed to fulfil the IEC 60601 regulations for electrical safety. The voltage generated in the connected impedance is measured by an instrumentation amplifier. The output voltage from the instrumentation amplifier is converted to a current signal that is fed to the input of the AD5933.

The general structure of the Analog Front End is composed of a high-pass filter to remove the DC voltage bias; a voltage to current converter, the front end converts the voltage output of the AD5933 to current that will be injected to the patient's body; a differential amplifier to sense the voltage through the receiving electrodes and minimize the load effect in the electrodes thanks to the high impedance in its input pins; and finally a voltage to current converter that returns the signal measured to the AD5933 and adds the DC component removed at the beginning in the first part.

To avoid potential health hazards the AD5933 and the front end are electrically decoupled using optocoupler technology from the rest of the board, this way power surges of any kind will not affect the patient in the case the board is connected to a power supply.

3.1.3 Microcontroller

The main core of the system is a Microchip PIC24FJ256GB106 microcontroller [18] with 256Kbytes of Program Memory, with up to 4 Universal Asynchronous Receiver-Transmitter (UART) modules, 3 Serial Peripheral Interface modules, 3 Inter Integrated Chip (I²C) modules and 16 channels input channels for the 10-bit analog-to-digital module.

The microcontroller is fed with an 8 MHz clock signal but is clocked at 32MHz and features low consumption (1 mA/MIPS, 2.0V Typical) with special modes of operation (Idle and Sleep) that further lower power consumption when the system is required to power down.

3.1.4 Bluetooth Communications

The board sports Bluetooth Communications using the Routing Networks RN-41-SM Bluetooth [19] socked module. Being a Class 1 Bluetooth device has a range of approximately 100 meters. This device is capable of operating with a baud rate up to 921Kbps using the Serial Protocol Profile (SPP) Bluetooth profile which emulates the behavior of a RS232 serial com port. The module can be programmed to require a password to pair up with the device and offers a myriad of power saving settings which can be configured.

3.1.5 Power Module and Battery charger.

The battery used in this board is a standard LiPo single cell battery supplying 3.7V and with a capacity of 750 mA/h. To charge the battery the board uses the 5V power supply from the USB bus. The feature of charging the battery using the USB port has become a standard thanks to the simplicity of the USB interface and the abundance of devices that can be used to supply power to the board.

3.2 Microchip MPLAB v8

The first objective in this thesis is the development of a firmware for the microcontroller PIC24FJ256GB106 to implement the synchronous impedance measurements using the AD5933-AFE and transmit the data using the Bluetooth channel. For that end Microchip® has a full development suite and hardware consisting on: MPLAB IDE, C30 compiler and ICD2 programmer.

The MPLAB IDE is a unified IDE (Integrated Development Environment) that eases the tasks of writing source code by coloring the code, compiling and building the project by adding automatically the parameters to the compiler and the debugging process by managing breakpoints and allowing peeking into (watch) the values of variables and processor registers.

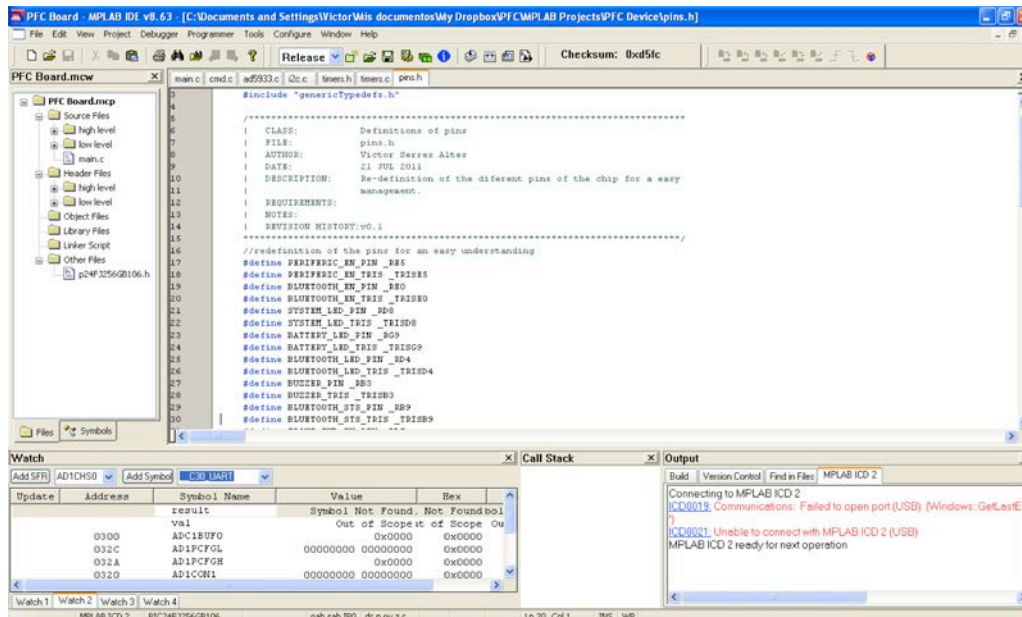


Figure 6 MPLAB IDE Screenshot

Microchip offers the MPLAB C30 compiler to allow the programming of their chips using C language. The C30 compiler is distributed using an Evaluation License free of charge. In this kind of license the advanced features of the compiler are only available the first 30 days of use.

In order to program the chip and help in debugging tasks Microchip® chips require the use of an ICD (In-Chip Debugger) like the ICD2. The ICD [20] and the chip communicate through the In Circuit Serial Programming (ICSP) protocol; the ICD uses a 4 pin bus to connect to a pair of dedicated pins, PGEDx and PGECx; the two extra pins in the bus are the power lines. Using the ICSP bus and a proprietary firmware in the chip the ICD and the MPLAB IDE provide means to debug the software using breakpoints, register and variable watches among other tools.

3.3 Microsoft Visual Studio 2008

The other objective of this thesis is to provide with a user-friendly interface to capture and display the obtained data received from the board through the Bluetooth serial port. The application's main feature is a graphical plot of the received data and the ability to export the data to a plain text file. The graphical area allows overseeing the data both while collecting and after the measurement process. The option of exporting the data collected to a file allows posterior reviewing of the data with the same application or processing the data with third party tools.

The application also allows the user to provide the AD5933 the calibration impedance value. A single impedance measurement command can be issued to check the correct status of the measurement equipment.

3.3.1 Zedgraph Open Source Plot Library

Visual Basic natively does not offer any ActiveX GraphControl to plot and display easily real-time changing graph data. To alleviate that problem the UI will use the Open Source Library ZedGraph Control [21].

ZedGraph is a powerful and free tool capable of drawing easily all kinds of 2D line, bars and pie charts. It is well documented, includes already implemented multitude of basic functions like zooming, panning, legend management, auto scale and much more and can be downloaded free from sourceforge.net.

3.4 Impedance Measuring Unit Validation Tests

Two type of test will be performed to validate the ICG measuring system. The first test will be done to determinate the impedance maximum frequency estimation of the AD5933 chip that will set up the maximum impedance frequency estimation. The second test will validate the performance of the ICG unit performing ICG thoracic measurements.

3.4.1 ICG Unit Maximum Impedance Estimation Sample Frequency tests

To validate the AD5933 maximum sampling speed a series of timed measurements will be made. These measurements will serve to observe the measurement cycle time. To time the cycle a timer of the PIC24 will be used. When fed with a 32 MHz clock the PIC24 timers are capable of having a precision up to hundredths of second.

3.4.2 Thoracic Impedance Measurements

When ordering a new measurement an on-board LED will be switched on. The system will continuously poll the AD5933 status bit to check if the measurement has ended and switch off the LED when the measurement ends. The switch time will be measured using a Tektronik MSO2012 oscilloscope. Transthoracic Impedance Measurements

The impedance system will be validated using two series of measurements using both textile and repositionable non-textile electrodes on two young adult males. Both series will use the same tetrapolar configuration as explained in the chapter 2 which is two electrodes in the neck and two more electrodes positioned at the base of the abdomen.

The AD5933 settings will be the lowest possible sampling time with 50kHz AD5933 output frequency. The AD5933 also requires setting an increase frequency value but as the measurement will be done on a single frequency the increase frequency register will have a value of 1kHz.

The electrode type choice has been to use both textile and repositionable non-textile electrodes. The repositionable 3M EKG Ag/AgCl electrodes [22] offer low impedance and are, in overall, like any other Ag/AgCl considered the best electrodes for biological and medical applications [8]. With their size of 12.58 cm² and their repositionable nature allow for placement corrections between measurements.

The textrodes created by Juan Carlos Marquez at the School of Engineering at University of Borås are made of synthetic wrap knitted fabric. The textrodes have a contact area of approximately 13.5 cm^2 made of conductive Velcro with loops of silver coated fibers.

After the measurements have been obtained they will be processed using a MATLAB script specifically designed to obtain a usable ICG signal, its frequency components and the extracted respiration signal.

The script applies a band pass filter with $\frac{2}{F_s}$ and $\frac{16}{F_s}$ band pass frequencies being F_s the measurement sampling frequency and a stop band filter between frequencies $\frac{11.6}{F_s}$ and $\frac{11.8}{F_s}$. The respiration signal is obtained applying a low-pass filter with a stop frequency of $\frac{2}{F_s}$. The frequency components of the ICG signal are obtained by calculating the fast Fourier transformation of both the processed and the unprocessed ICG signal.

This chapter will present the results of the work done in the thesis. The information here presented focuses primarily on the results obtained while the difficulties encountered during the developing process will be dealt in the next chapter.

4.1 Measuring Unit Firmware

As described in the previous chapter the main motivation in this thesis is the implementation of the firmware that can fulfill the timing requirements of continuous impedance measurement.

The resulting firmware offers a command line interface (CLI) accessible by the two serial interfaces of the board, Bluetooth and USB, and accepts written commands similarly to a Unix-like terminal. The commands accepted are discussed in a later section.

4.1.1 Structure of the project

The source code of this project is organized in different libraries and files that, with a little extra work, could be separated into different modules which could be easily integrated into another projects based in the same board. In fact the libraries controlling both the AD5933 chip and the I²C communications bus are used in another bachelor thesis conducted by Juan García under the tutelage of Javier Ferreira.

The resources of the PIC24 used up by this project are:

- 2 of the 5 16-bit timers
- 2 of the 4 UARTs
- 1 of the 3 I²C modules

The two timers are used for different purposes. The timer 1 of the PIC24 is programmed to tick every millisecond and simply increment/decrement several variables on which many processes depend on like delays in the code or timeouts while the timer 2, which is configured to tick as the sampling time required for the system is the one issue the AD5933 the repeat frequency command, more on this later.

Generally speaking the firmware has three modes of operation: Sleep mode which will be described in the power management section, command mode and measurement mode.

The default mode of operation is the considered the command mode and it's the mode that the PIC will return in case there is any problem with the communications. In this mode the PIC waits for any command to be received on any of the two UARTs and once received processes it.

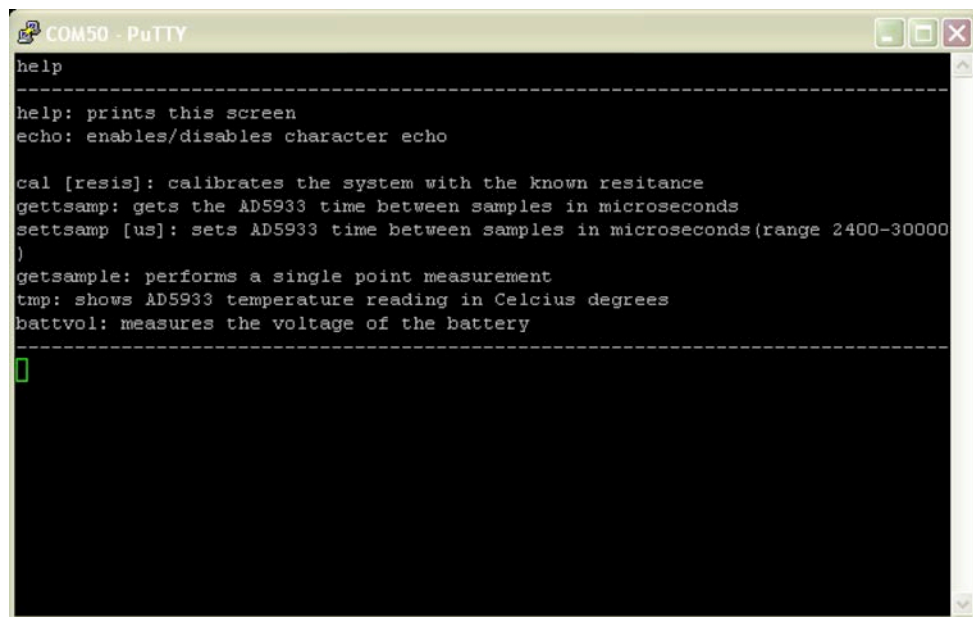
Once a certain command has been issued the firmware enters in measurement mode where starts measuring continuously impedance using the AD5933 and sending the information to the computer to be received by the UI application while ignoring most commands trying to minimize interference.

To avoid blocking the firmware in measurement mode the firmware reverts to command mode if no feedback has been received from the computer while sending data once in a few seconds.

4.1.2 Commands

The firmware has a Command Line Interface (CLI) and accepts commands consisting by a string ended by a carriage return character ‘\r’ of one or two words separated by a space character. Both the command and the argument must not exceed 10 characters in length. The number of arguments accepted can be changed by modifying the *MAX_ARGS* constant in the *cmd.h* file. Below there is a list of the available commands:

- *help*: prints the help screen detailing the functions available to the user and their function. See figure 1.



```
COM50 - PuTTY
help
-----
help: prints this screen
echo: enables/disables character echo

cal [resis]: calibrates the system with the known resitance
gettsamp: gets the AD5933 time between samples in microseconds
settsamp [us]: sets AD5933 time between samples in microseconds(range 2400-30000
)
getsample: performs a single point measurement
tmp: shows AD5933 temperature reading in Celcius degrees
battvol: measures the voltage of the battery
-----
█
```

Figure 7 Help info displayed

- *echo*: enables the character echo. This command is intended to be used as a help for a human user that accesses the system via a terminal application and as such it has the opposite effect of the *cmp* command on identifying the user.
- *cal [resis]*: calibrate command. The value of the known impedance must be sent as the argument. The PIC expects the user to send the value of the known impedance as a string if operating using a terminal or as an IEEE float in binary form (4 bytes) if using the application.
- *gettsamp*: sends the sampling time in milliseconds in string format

- *settsamp [time in us]*: configures the sampling time. The argument must be the time in microseconds. The values accepted range from 2400 to 30000.
- *getsample*: performs a single point measurement. If the user is accessing the board via the terminal the data is returned as a string representing the decimal value with 2 decimals. On the other hand, if the user is using the application the value is sent as an IEEE float in binary form.
- *tmp*: the AD5933 is capable of measuring its own temperature. This function returns the temperature of AD5933 in Celcius degrees. The typical values are 27°C if the output signal of the AD5933 is turned off and the board is at room temperature; 31°C is the typical value if the AD5933 operating and is injecting current to an impedance.
- *battvol*: gives an approximate value of the voltage of the battery, it should be enough to determine if the battery has charge or it's near depletion.
- *cmp*: informs the board that the user is using the windows application and turns off the serial echo if activated. It also works as a simple mechanism to ensure that there is communication between the computer and the board as the PIC answers with an "OK" every time it receives a *cmp* command in command mode.
- *startm*: initiates the procedure to start measuring continuously, after some handshaking between the application and the PIC the PIC will start sending the measurement data continuously while it receives a *cmp* command once in at most 5 seconds and until an *stopm* command has been received
- *stopm*: stops the measurement process and returns the PIC to command mode.
- *adrestart*: it is a simple way to cause a hard-reset to the AD5933 by turning off the power supply to the isolated area and turning it on shortly after. The command was added as a debug tool and, while it is still working, it is not intended for regular use.

4.1.3 Power management

The board is designed to work with a battery and because of that the power drain is a concern. To minimize the power waste the firmware enters the system in sleep mode after 1 minute without receiving commands though the Bluetooth UART in command mode.

Sleep mode is a special mode of the PIC24 series of chips that has the following features[18]:

- The system clock source is shut down. If an on-chip oscillator is used, it is turned off.
- The device current consumption will be reduced to a minimum provided that no I/O pin is sourcing current.

Before entering sleep mode the firmware turns off the power to the peripherals and the electrically isolated area: frontend and AD5933.

The firmware is programmed to wake up the PIC when the switch on button is pressed. After the PIC has awoken-up the firmware will restore power to the peripherals and to the AD5933.

The PIC can also be forced to enter sleep mode by pressing during 2 seconds the switch-on button.

4.1.4 Measured Data Protocol.

On measurement mode the PIC sends the measured impedance value through a serial port to a computer continuously with a format similar to a data packet similar to other protocols. Figure 2 represents the format of an impedance data packet.

SYNC_BYTE (1B)	Impedance data (4B)	CRC (1B)
-------------------	------------------------	-------------

Figure 8 Data packet format

The first byte sent is a byte of synchronization. Due to being wireless one of the serial interfaces some communication problems can be expected due to heavy interferences or excessive length of the link. The windows application always expects having this byte before reading the impedance data so in case of lost or corrupt data the application will discard all the packets until this byte is read. By default the value of this byte is 66 in decimal or 0b01000010.

The impedance data are 4 bytes that represent an IEEE-754 floating point number.

The Cyclic Redundancy Check (CRC) byte is an error detection byte to check if the impedance datum received is valid. In this thesis the operation to calculate the CRC is rather simple and does not follow the industrial standards commonly found in other projects or communication standards. The CRC value is simply the sum of all the 1's in the synchronization byte and in the 4 bytes of data.

4.1.5 AD5933 and frontend configuration

As this thesis is focused on single frequency impedance measurements the frequency at which the AD5933 measures is 50 kHz which is the *de facto* standard used in the scientific community. In order to keep to a minimum the delay between the order of measuring frequency and the actual start of the measurement process it has been chosen to have 3 setting cycles at the beginning of every impedance measurement.

To calibrate the system each time before measuring or after a restart a 100Ω 250mW carbon resistor was used. The frontend was originally designed for a full body measurement configuration. In order to realize measurements of impedance cardiography and adapt the reception instrumentation amplifier to the dynamic range of the thorax impedance (which ranges approximately from 25Ω to 45Ω) the front-end voltage instrumentation amplifier gain resistor was changed for a 2600Ω resistor.

4.1.6 Critical aspects of for a successful operation

Impedance measurement range

The AD5933 has available a highly precise 12-bit Analog-to-Digital Converter (ADC) but for it to make precise conversions the dynamic range of the impedance cardiography measurement, i.e. the impedance of the thorax, must be adapted to the conversion range of the ADC. To successfully achieve that one resistor of the analog front-end original designed for total right side measurements must be changed. This fact renders the device unreliable making measurements of other parts of the body. While it's not a critical aspect of operation it affects the versatility of the device and it's noteworthy.

Sample Time

In order to be able to produce an accurate analysis on the measured cardiogenic impedance, measurements must be taken with enough time resolution. The performed test on the implemented system indicate that the minimum sample time is 2.4 ms, which produced a maximum sample frequency of 416 Hz. Measuring Impedance cardiography with such time resolution is accurate enough to perform any analysis of any cardiogenic activity.

4.2 User Interface application

As described in chapter 3 the application developed for Windows PC using Visual Basic 2008 features a plot area and allows exporting the measured data to a file.

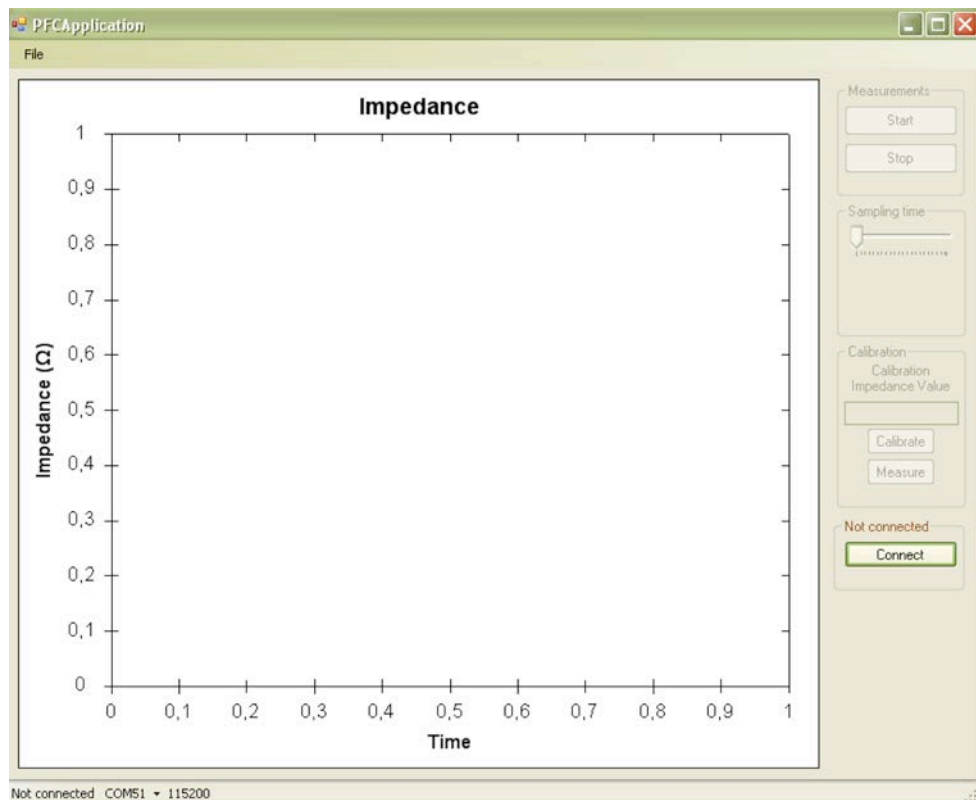


Figure 9 Application Screenshot

4.2.1 Status bar

The status bar is located at the bottom of the form and shows the connection status, Not Connected or Connected, the current COM port in use and the serial port speed in bauds per second. Figures 3 and 4 show examples of information displayed by the status bar.

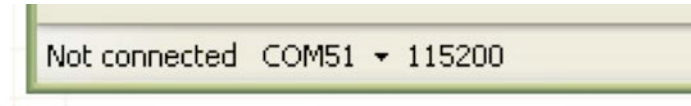


Figure 10 Status bar disconnected

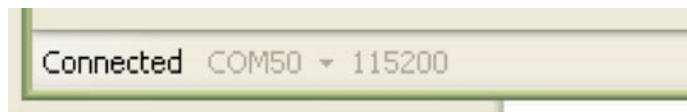


Figure 11 Status bar connected

The default speed is set to 115200 bauds per second but it can set to any value that the serial port drivers admit. Please note that to ensure that the measurement process works properly the baud rate should be of at least 20000bps for the fastest sampling rate and 1600bps for the slowest sampling rate.

The status bar also displays a message if recently there has been a problem connecting to the board. There are 5 situations where a timeout and/or a connection drop can be detected:

- While connecting to the board. There might be a host of problems while connecting to the board, starting by serial interface driver problems or the board being off. Every single problem will give a different error message.
- While setting the measurement parameters or calibrating.
- While checking if the connection is alive.
- While starting or stopping the continuous impedance measurement.
- While receiving impedance data.

A connection problem in any on the aforementioned situations will pop up a message box indicating the error. Only in the case of setting the parameters or calibrating and starting the measurement process the application will allow the user to continue to use the application without having to reconnect to the board; in the last cases the application will assume that simply there has been a delay in the communication.

4.2.2 Control area

The control area is divided in several sections each grouping the controls that control the same aspect of the board: Measurements, Sampling time, Calibration and Connect/Disconnect.

- *Measurements*: This group box houses the two buttons used for sending the *startm* and *stopm* commands.
- *Sampling time*: The sampling bar allows configuring the sampling time with a resolution of tenth of millisecond and a value ranging from 2.4ms to 30ms.
- *Calibration*: The calibration group box contains the textbox and button for calibrating the AD5933 and a button to perform a single point measurement, ideal to validate the calibration.
- *Connect/Disconnect*: The “connect” and “disconnect” buttons allow managing the connection to the board and will hide when unavailable.

4.2.3 Plot area

The plot area is the biggest part of the UI of the program and is the primary feature of the windows application.

By default the graph resolution sets itself to show the samples collected during the last 3 seconds to avoid overwhelming the user with data. To change the number of samples displayed there are two buttons and a textbox below the plot area.

When the number of samples collected is greater than the ZedGraph control is configured to show a scrollbar appears to allow the user to navigate through the whole data.

4.2.4 File system

The data measured can be saved on a file and be reviewed later using the same application or import it to another program like MATLAB.

The data is saved in ASCII format in a plain text file with a *.txt* extension. The first line of the file is reserved to the exact date, up to tens of microseconds using the computer local time, of the beginning of the measurement process using the computer clock. In the second line the sampling time is recorded in microseconds. After the second line each line has a value of impedance with 3 decimal positions maximum. Figure 5 shows as an example the first 6 lines of a file that was generated when conducting the final tests.

```
Timestamp: 07/21/2011 12:15:04.82812
Sampling time 2400
43,6645431518555
42,9368324279785
43,7164726257324
43,7445106506348
[...]
```

Figure 12 Example of text file

By default, the file is saved in the Documents folder of the windows actual user with the time of creation of the file as name but it can be changed using the *Save as...* option.

4.3 Validation Measurements

4.3.1 Timing requirements

As written in chapter 3, the AD5933 underwent a series of tests to time the minimum sampling time.

Before starting the baud rate of the I2C bus was measured to check if it complied the 400 kHz and adjusted the configuration accordingly. The AD5933 maximum output frequency was also measured and it resulted 500 kHz being much higher than specified in the datasheet [17]. With the help of timers the correct speed of the PIC24 internal clock was also assessed. The internal PLL correctly sped up the 8 MHz clock fed to the PIC24 to 32 MHz.

The minimum sampling time recorded was 2.4ms, which corresponds to a frequency of 416 Hz; any value below that minimum would not behave consistently after a large number of samples.

4.3.2 ICG Measurements results

The ICG signal was successfully measured using both types of electrodes. This section will present the results obtained:

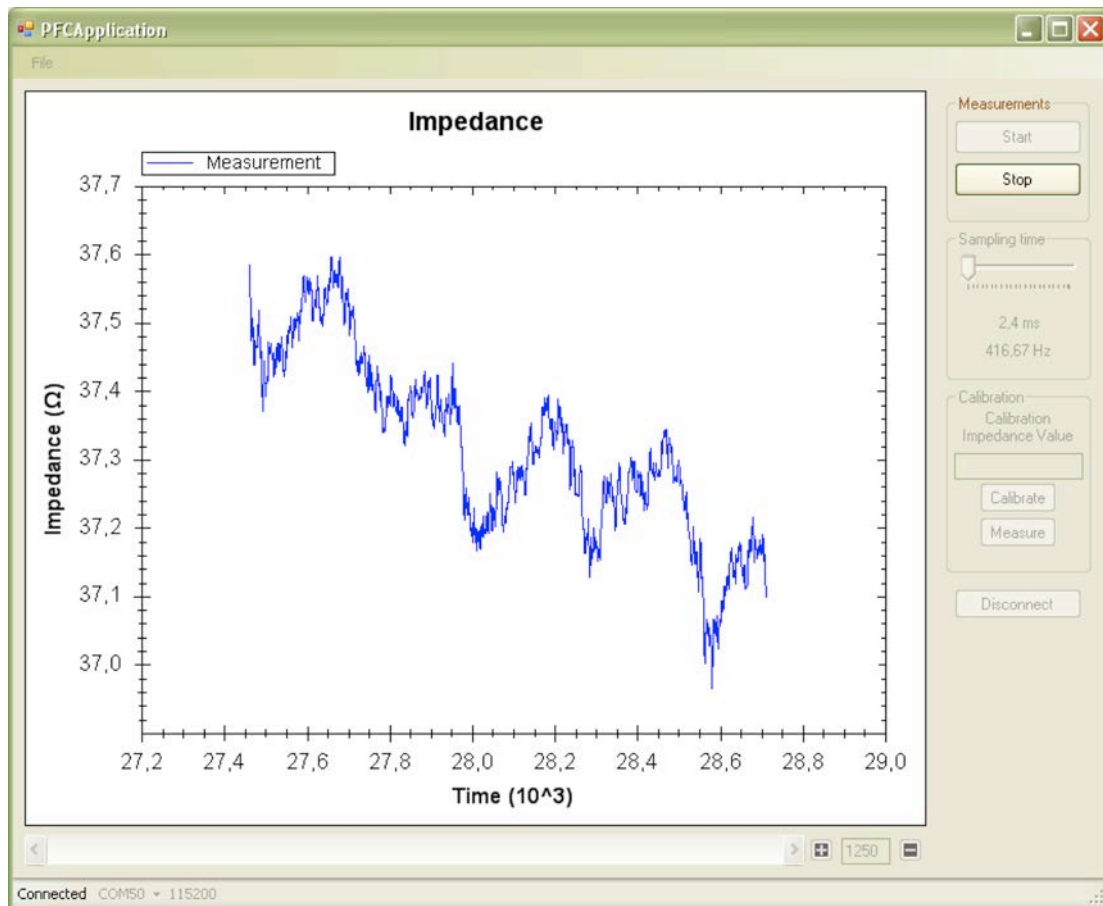


Figure 13 Application while measuring

Figure 7 shows a screenshot during the measurement process of one of the tests while Figure 8 shows a screenshot of a composite plot created by a MATLAB script combining three plots:

- The blue graph shows the raw impedance data extracted from the result file. This data is noisy and almost no interpretation can be extracted from it.
- The green plot shows the effect of breathing over the transthoracic impedance (TEB). The value of the impedance measured increases above or decreases below the baseline. Breathing in increases the volume of air in the chest region and increases the overall impedance while breathing out decreases it. Although these two facts are very well correlated they are not directly related, as explained in [23].
- The black plot is the interesting value of the impedance variation as a result of the changes of fluid volume in the heart.

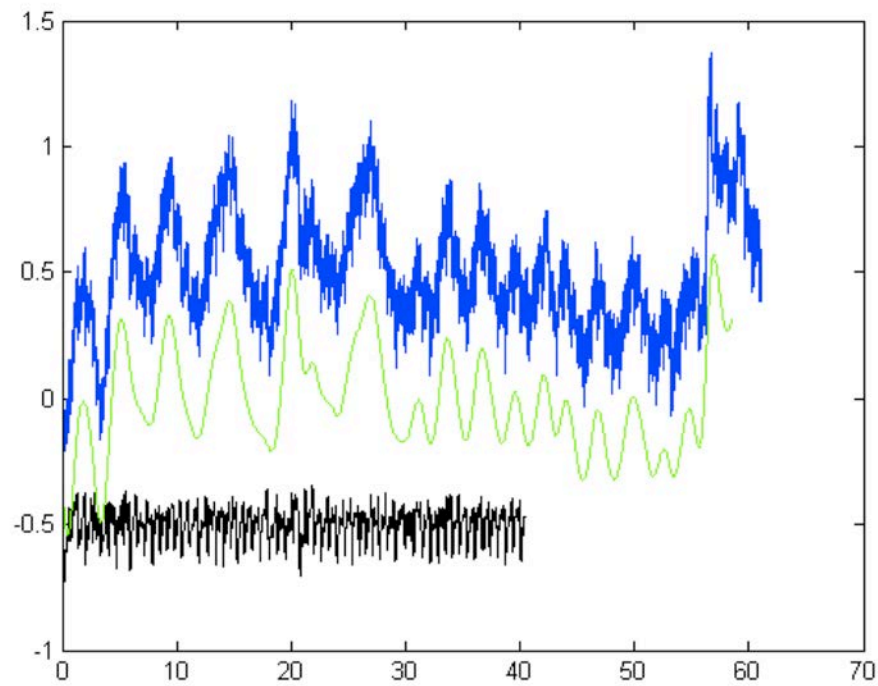


Figure 14 MATLAB Script Composite Plot

Both sets of test measurements were done to a young adult and standing up. The testing conditions weren't optimal as the testing subject didn't remain quiet and played a little with his breathing. The results show the expected ICG signal but with a heavy baseline distortion due to the effect of breathing.

The first measurement was done at 416 Hz with textile electrodes. Figures 8, 9 and 10 show the raw data, the filtered data and the first derivate of the filtered data of the textrode measurement respectively.

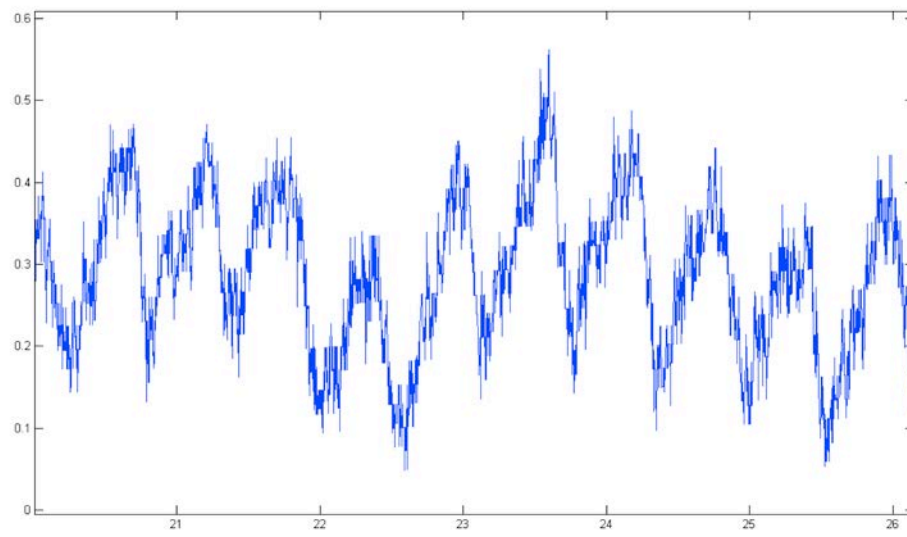


Figure 15 Raw data from the textrodes measurement

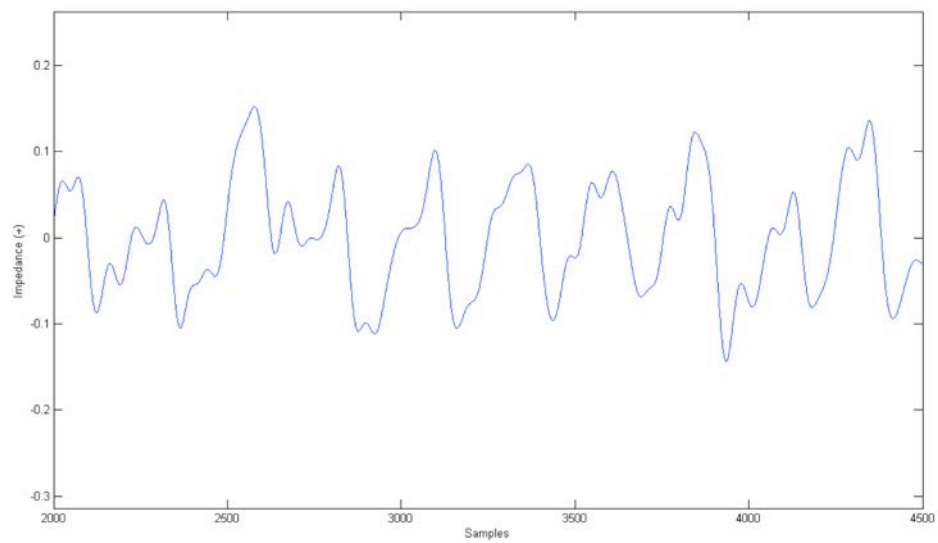


Figure 16 Filtered ICG data from the textrodes measurement

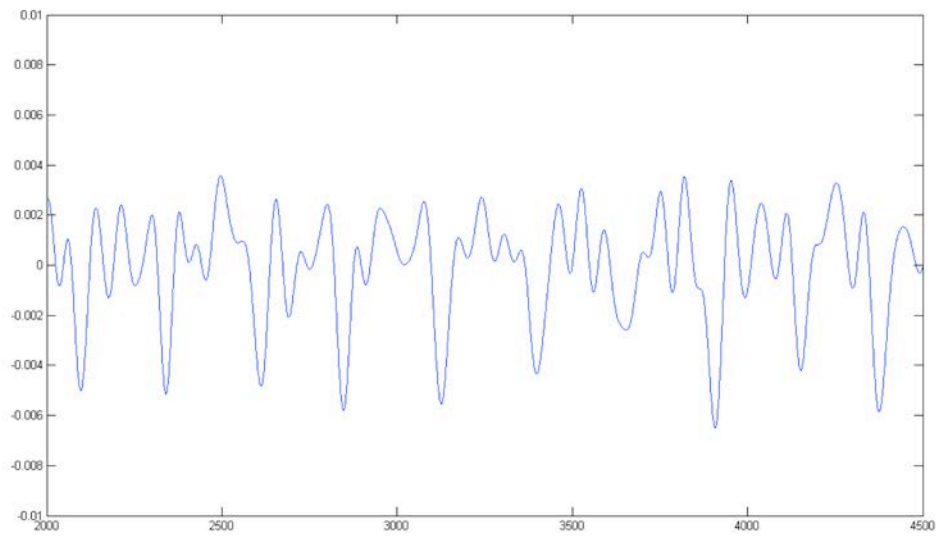


Figure 17 First derivate of the filtered tetrode measurement signal

The second measurement was done with the same sampling speed but using repositionable electrodes. Figures 11, 12 and 13 show the raw data, the filtered data and the first derivate of the filtered data respectively of the measurement with repositionable electrodes.

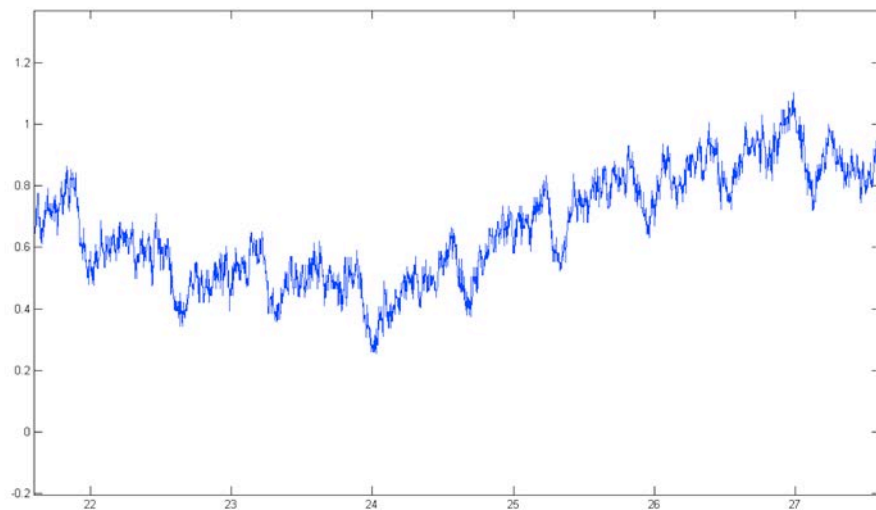


Figure 18 Raw data from the repositionable electrodes measurement

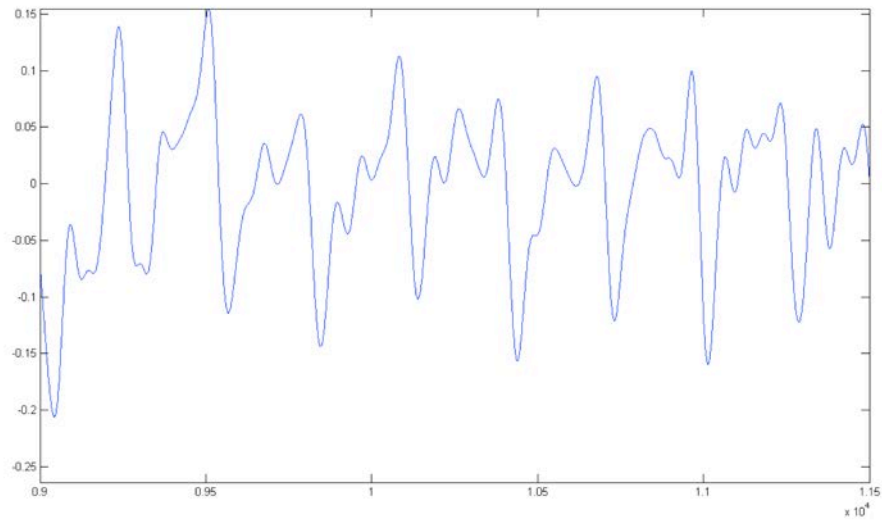


Figure 19 Filtered ICG signal from the repositionable electrode measurement

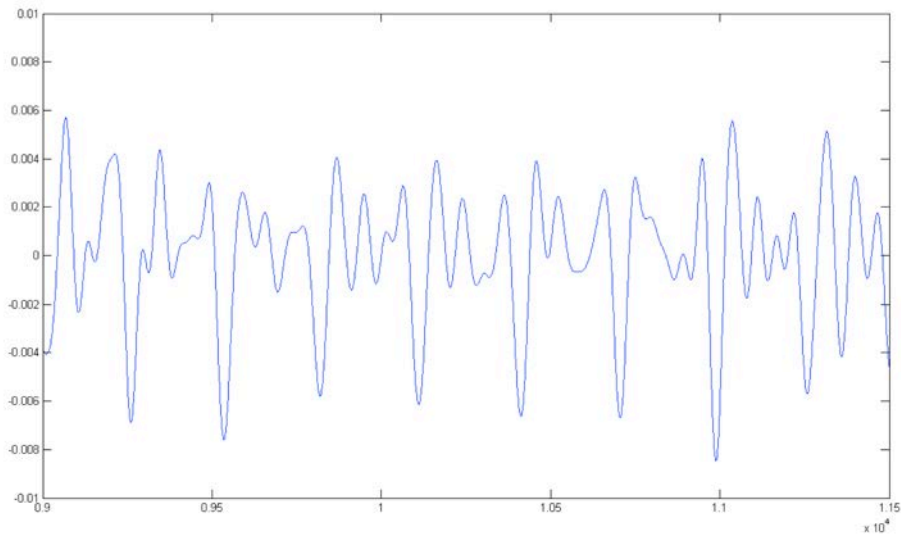


Figure 20 First derivate of the filtered ICG signal of the repositionable electrode measurement

Both figures 9 and 12 show the raw data, due to noise and the effects of artifacts no useful conclusions can be drawn from them. The noisy result of the raw data is typical of the measurements and can be easily filtered out.

Figures 10, 11, 13 and 14 show a periodic signal with period being the heart cycle. The impedance signal shows non periodicities due to the sensibility of ICG to motion artifacts [6].

Both measurements show correctly the data expected from a healthy adult male. The major difference between the two raw signals comes from the breathing effect as the test

subject played with his own breathing during the measurements and didn't refrain from talking.

Figures 11 and 14 show the first derivate of the ICG signal. The MATLAB script calculates it because in practice only the EKG signal and the first derivate of the ICG signal are used for calculating hemodynamic parameters [6].

Originally the of the measurement electronic system used for this thesis was designed with the purpose of measuring total body impedance to detect the presence of limb edema [9]. This thesis pushes the specifications of the original measurement system and its core, the AD5933 integrated circuit, in order to check if the board is suitable for continuous impedance measurements.

5.1 Transthoracic Impedance Measurements and Impedance Cardiography Analysis

The measurement range has been adjusted from the total body load to the thorax. This way the magnitude resolution is between 20Ω and 60Ω , with the 12-bit ADC of the AD5933, allowing the execution of an accurate wave analysis.

In order to perform any cardiography analysis through impedance, it is required a time resolution above 250 for the impedance measurement. In this work the maximum impedance sampling frequency of the AD5933 has been evaluated and the result indicates that impedance can be acquired at over 400 SPS. This confirms the AD5933 can be use indeed for ICG recordings.

5.2 Measurement Noise

Since the device is powered with batteries, interference from the means is not expected to be a mayor problem since in any case any 50 Hz eventually noise couple to the measurements leads can be easily remove through filtering.

5.2.1 Effects of Breathing on the Measurements

As shown in the graphs plotting raw impedance there is a denoted influence of the respiration activity in the TEB measurement.

Respiration activity influences the TEB measurement by modulating the ICG around its base value Z_0 . This effect, being of low frequency, is best eliminated by digitally filtering the ICG signal [6].

CONCLUSION AND FUTURE WORK

6.1 Conclusion

The conclusion that can be extracted from the test results is that has been shown that the AD5933 is suitable for use in bioimpedance measurement applications that have a requirement of continuous sample acquisition e.g. impedance cardiography measurements and respiration monitoring. Being at 2,4ms, the maximum sample rate achieved with the AD5933 is high enough to be used in applications demanding a high temporal resolution.

6.2 Future work

The work of this thesis has been shown to be able to operate successfully with textile electrodes. This opens possibilities of developing clothing with incorporated electrodes which would allow the development of a cardiac holter-type with bioimpedance measurements for athletes.

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