Performance of an implantable impedance spectroscopy monitor using ZigBee

P Bogónez-Franco¹, A Bayés-Genís², J Rosell¹, R Bragós¹

¹ Electronic Engineering Department. Universitat Politècnica de Catalunya. Campus Nord. Edifici C4. C. Jordi Girona 1-3. 08034 Barcelona. Spain.
² Cardiology Service. Hospital de la Santa Creu i Sant Pau. C. Sant Antoni Maria Claret 167. 08025 Barcelona. Spain.

E-mail: bogonez@eel.upc.edu

Abstract. This paper presents the characterization measurements of an implantable bioimpedance monitor with ZigBee. Such measurements are done over RC networks, performing short and long-term measurements, with and without mismatch in electrodes and varying the temperature and the RF range. The bioimpedance monitor will be used in organ monitoring through electrical impedance spectroscopy in the 100 Hz - 200 kHz range. The specific application is the study of the viability and evolution of engineered tissue in cardiac regeneration in an experimental protocol with pig models. The bioimpedance monitor includes a ZigBee transceiver to transmit the measured data outside the animal chest. The bioimpedance monitor is based in the 12 Bit Impedance Converter and Network Analyzer AD5933, improved with an analog front-end that implements a 4-electrode measurement structure and allows to measure small impedances. In the debugging prototype, the system autonomy exceeds 1 month when a 14 frequencies impedance spectrum is acquired every 5 minutes. The receiver side consists of a ZigBee transceiver connected to a PC to process the received data. In the current implementation, the effective range of the RF link was of a few centimeters, then needing a range extender placed close to the animal. We have increased it by using an antenna with higher gain. Basic errors in the phantom circuit parameters estimation after model fitting are below 1%.

1. Introduction

Electrical impedance monitoring is a valuable technique used to monitor the tissue and organ state and electrical impedance spectroscopy (EIS) has been used for many years for cardiac tissue characterization to detect acute ischemia [1], [2], old infarction scar [3] and rejection in transplanted heart [4], [5]. Measurements are performed in open thorax set-ups [2], through catheter [2], [3] or using a pacemaker-type device [4].

We are now working with the Cardiology Department of Santa Creu i Sant Pau Hospital in non-invasive graft rejection detection and regenerative tissue engineering. In the last case a non-functional myocardium area is intended to regenerate by using myocardial cell precursors derived from stem cells. In order to monitor the viability and evolution of the engineered tissue after its implantation in an animal model heart (pig or sheep), an implantable device should be developed. In this paper we present the preliminary results of a device which is capable to acquire an impedance spectrum every 5
minutes in the range of 100 Hz – 200 kHz during 1 month [6]. Some characterization measurements
are presented.

2. Materials and methods

2.1. Bioimpedance monitor
The bioimpedance monitor developed [6] measures the electrical impedance at 14 frequencies, stores
the data collected and transmits them to an external computer using a ZigBee protocol [7]. The system
has a size of 35 mm x 37 mm x 8 mm. The bioimpedance monitor is powered from a NiMH battery.
The system makes a measurement every 5 minutes and remains the most part of time in sleep mode,
increasing the autonomy. When the system is in sleep mode the current consumption is 0.2 mA and
when it is doing a measurement the current consumption is 35 mA. The bioimpedance monitor is
designed to measure in the range 10Ω to 1 kΩ, at a programmable set of frequencies.
The connection of the bioimpedance monitor with the needle electrodes is done using short and thin
coaxial cables. Figure 1 shows a system diagram.

2.2. Measurements
All characterization measurements were done using a series RC network with the following values, Rs
= 68 Ω, Rp = 130 Ω and C = 100 nF, and over a 330 Ω resistor and with the electrode impedance
network attached. The electrode impedance network was modeled as a parallel RC network with the
following values, Rp = 10 kΩ, Rs = 240 Ω and Cs = 18 nF. The measurements performed were short-
term, medium-term, with and without electrode mismatch and temperature changes. The temperature
change was between 25 ºC and 40 ºC results. For long-term measurements are shown in [6].

3. Results

3.1. Short-term and mid-term measurements
The short-term measurements where done over a period of time of 5 minutes. With this measurement
we would like to see the effect of drifts during the warm-up time. The medium-term measurements
where done over a period of time of 5 hours. Figure 2 shows the error at each frequency when the RC
network is measured.

Figure 1: System diagram.

Figure 2: Error at each frequency when a RC network is measured.
3.2. Electrode mismatch

To see the effect of electrode mismatch in the measurement we produce an increment of 100% of the electrode impedance in each lead of the impedance monitor. Electrode mismatch is a very common problem in all impedance measurements and is caused by a poor contact between the electrode and the tissue under measurement. Normal needle electrode impedance is in the range of 10 kΩ or more at low frequencies, whereas at high frequencies its value is around 100 Ω. When the electrode mismatch is produced in the injecting leads, the error is in the order of 3.5% and has periodicities at each decade. When the electrode mismatch is produced in the sensing leads the effect is negligible, less than 1%.

Temperature profile
We measure the temperature effect in module and phase measuring a 330 Ω resistor. The profile consists in three steps of an increment of 5 ºC each, see figure 3.

![Figure 3: Variations in module and phase over a 330 Ω ohms resistor to a temperature change.](image)

3.3. RF Range

The range measurements were done with the impedance monitor enclosed in a plastic container and immersed into a fishbowl (30 cm x 15 cm x 15 cm) filled with saline solution with a conductivity of 16 mS/cm. As we exposed in a previous work [6] the plastic container was moved at different locations inside the fishbowl to measure the distance when the RF link was broken. In order to increase the range we changed the SMD antenna of the Access Point (AP) for a 5-element Yagi antenna. The figure 4 shows the RF range using the Yagi antenna.

![Figure 4: Graph of the impedance monitor RF range. The x-axis corresponds to the longitudinal distance of the plastic container from the border of the fishbowl. The y-axis corresponds to the distance at which we measure the range. The z-axis corresponds to the depth of the plastic container inside the fishbowl.](image)
4. Discussion

The change in the common mode voltage induced by an electrode mismatch in the injecting leads, combined with the output impedance of the current source, has a higher effect in the measurement than the same mismatch produced in the sensing leads, due to the high impedance of the input stage.

The most important effect of temperature changes is due to the slope of the profile than the increment itself. As the temperature remains stable the value of the module returns to its original value. This may be produced by the heterogeneous warming of the internal integrated circuits causing the in-balance in current sources. The phase is less sensitive to temperature variations.

Adding an antenna optimized for the ZigBee band the range can be increased 50% the worst case. of the RF link. This could allow us to remove the range extender (RE) that we have planned to use in the outer surface of the animal.

Acknowledgements

This work has been supported by grants from Spanish Ministry of Science and Innovation SAF 2008-05144, with FEDER funds, and from the Spanish Ministry of Health project REDINSCOR.

References

[1] Gebhard M M, Gersing E, Brockhoff C J, Schnabel P A, Bretschneider H J 1987 Impedance spectroscopy: a method for surveillance of ischemia tolerance of the heart Thorac. Cardiovasc. Surg. 35 1 26-32
[2] Salazar Y, Bragos R, Casas O, Cinca J, Rosell J 2004 Transmural versus nontransmural in situ electrical impedance spectrum for healthy, ischemic, and healed myocardium IEEE Trans. Biomed. Eng. 51 8 1421-7
[3] Warren W, Bragós R, Casas O, Rodriguez-Sinovas A, Rosell J, Anivarro I, Cinca J 2000 Percutaneous electrocatheter technique for on-line detection of healed transmural myocardial infarction Pacing Clin. Electrophysiol. 23 8 1283-7
[4] Pfitzmann R, Müller J, Granhan O, Hetzer R 2000 Intramyocardial impedance measurements for diagnosis of acute cardiac allograft rejection Ann. Thorac. Surg. 70 2 527-32
[5] Cinca J, Ramos J, García M A, Bragós R, Bayés-Genís A, Salazar Y, Bordes R, Mirabet S, Padró J M, Picart J G, Viñolas X, Rosell-Ferrer 2008 Changes in myocardial electrical impedance in human heart graft rejection Eur. J. Heart Fail. 10 6 594-600.
[6] Bogónez-Franco P, Bragós R, Bayés-Genís A, Rosell J 2009 Implantable bioimpedance monitor using ZigBee Engineering in Medicine and Biology Society EMBC 2009 Annual International Conference of the IEEE 4868-71
[7] SimpliciTI 2009 www.ti.com