INTERNATIONAL CONGRESS ON COMPUTATIONAL BIOENGINEERING
M. Doblaré , M. Cerrolaza and H. Rodrigues (Eds.)

SYSTEM FOR MEASURING THE TRANSTHORACIC ELECTRICAL IMPEDANCE TO THE ECG SIGNAL

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ABSTRACT

This paper presents an electronic system for measuring the impedance of the electrocardiograph signal from its origin (heart) to the electrodes placed on the patient's skin. The main advantages of this system are that the reading is taken at the frequency of the electrocardiographic signal and that it is a totally non-invasive method, as no external current is applied to the patient besides the amplifier polarisation currents. The developed system allows the measurement of a physiological parameter related with anthropometric variables such as height and body surface area, and reflects pathological conditions that theoretically can alter thoracic impedance such as chronic bronchitis and thoracic surgery. Besides, it allows a correction of the ECG derived parameters that increases the correlation with left ventricular mass and improves the senstitivity of ECG for the diagnosis of left ventricular hypertrophy.

1. INTRODUCTION

One of the electrical variables for obtaining information on the existence of certain pathologies and the body composition of a patient is the electrical impedance. This is a non-invasive, low-cost method that enables reasonably precise diagnoses to be made. It is used, for example, in studies of body composition, measurement of the cardiac output, detection of a patient's respiration, image formation, etc.

The method used involves applying a high-frequency signal (1 Khz - 10 Mhz) and determining the equivalent electrical model, usually represented by an RC parallel circuit.

The impedance reading is twofold:

a) Z due to the electrode-electrolyte interface in the connection of the electrodes to the skin and

b) the patient's internal tissue impedance.

Curves representing the variation of impedance with frequency are well known. At low frequencies (close to 0 Hz.) impedance rises sharply, due mainly to the electrical model of the electrode-electrolyte interface, which clearly shows a high impedance at low frequencies.

Cardiography impedance is a valid technique for determining cardiac parameters: with 2 electrodes placed on around the neck and another 2 around the thorax, a low-amplitude, high-frequency signal is applied to obtain the impedance. The changes therein are due to variations in intravascular and extravascular fluids. The cardiac output represents the volume of blood expelled during a cardiac cycle.

Several equations for the calculation of stroke volume have been proposed. Based on the work of Nijboer [1], Kubicek et al. [2] used the next expresion:

$$SV = \rho \frac{L^2 T}{Z_0^2} \left(\frac{dZ}{dt}\right)_{MAX} \quad (1)$$

where SV is the ventricular volume; ρ is the blood's resistivity to the excitation frequency; L is the inter-electrode distance; Z_0 is the baseline rest impedance; T is the ventricular ejection time and dZ/dt is the variation rate of impedance. The value of ρ and L are not easily

This work has been supported by Ministerio de Ciencia y Tecnología (Spain): project SAF2000-0221-C02.

measurables for each patient so Sramek et al. [3] modified the last expression as follows:

$$SV = \frac{0.00117 * h^3}{Z_0} \left(\frac{dZ}{dt}\right)_{MAX} \qquad (2)$$

where h is the patient´s height.

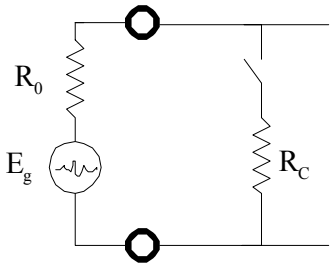

Fig. 1. Electrical model

The main problem is posed by the fact that the impedance varies with the patient's respiration and movements. The equation assumes that the thorax is a cylinder (frustoconical in eq. (2)) and that the relation between the impedance and flow is linear. Attempts to eliminate the height of the patient have been recently published [4].

In many cases the impedance was considered to be purely resistive whereas in fact it varies with the frequency.

In all cases the procedure is the same one: a current is applied among the electrodes and the voltage is measured. With these data it is possible to obtain the impedance value [5][6]. The waveforms of injected current can also differ: sinusoidal, impulses, steps, or more complexly shaped pulses of current, which contain different frequency components in a composite waveform, can be injected.

Little interest has been shown in measuring low frequency impedance. In the present case our interest lies in measuring the electrical impedance of the electrocardiographic signal from its generator (cardiac muscle) to the external amplifier of the extracellular signal. A simple method of circuit theory was used for making the measurement (Fig. 1). We considered the heart as an ideal generator (E_g) of serial signals with purely resistive impedance (R_0). It is possible to obtain the value of the resistance in series with the heart by measuring the tension in open circuit and placing a known resistance (R_C).

The main advantages of this method are:

i. It is non invasive as no current is applied to the patient

ii. The impedance is measured at the frequency of the electrocardiographic signal

2. MATERIALS AND METHODS

The system designed for measuring the transthoracic impedance of the electrocardiograph signal involves considering the heart as a generator of serial signals with a resistance R_0 (at the ECG signal frequency the capacitive and inductive effects are not considered). Using a system of electrodes, the signal's amplitude and energy are measured when the input impedance of the amplifier system is much greater that the transthoracic impedance; this value will be Eg, voltage of the ideal generator; a resistor (R_C) is then set up in parallel with the electrodes and ipso facto with the input of the amplification system; the signal's amplitude and energy are measured and it is possible to obtain R_0.

If E_{OPEN} is the energy obtained in open circuit and E_{CLOSE} is the energy obtained on having closed the circuit with a resistance R_C, is fulfilled that:

$$E_{CLOSE} = \frac{R_0}{R_0 + 2R_C} . E_{OPEN} \qquad (3)$$

This process has been optimized to obtain more accurate readings of the impedance value. A new system (hardware + software) has been designed (Fig. 2), formed by an Acquisition Board (AB) and a Portable Computer (PC), both connected by a RS232 protocol.

Basically, the AB´s function is to pick-up the ECG signal in the two conditions indicated previously and send it to the PC; the PC allows controlling the functioning of the AB: open/close the relay, off-set and

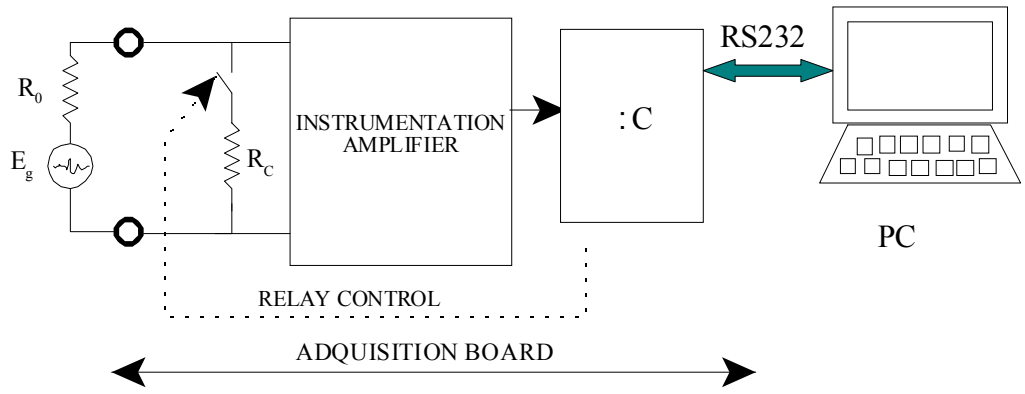

Fig. 2. System designed

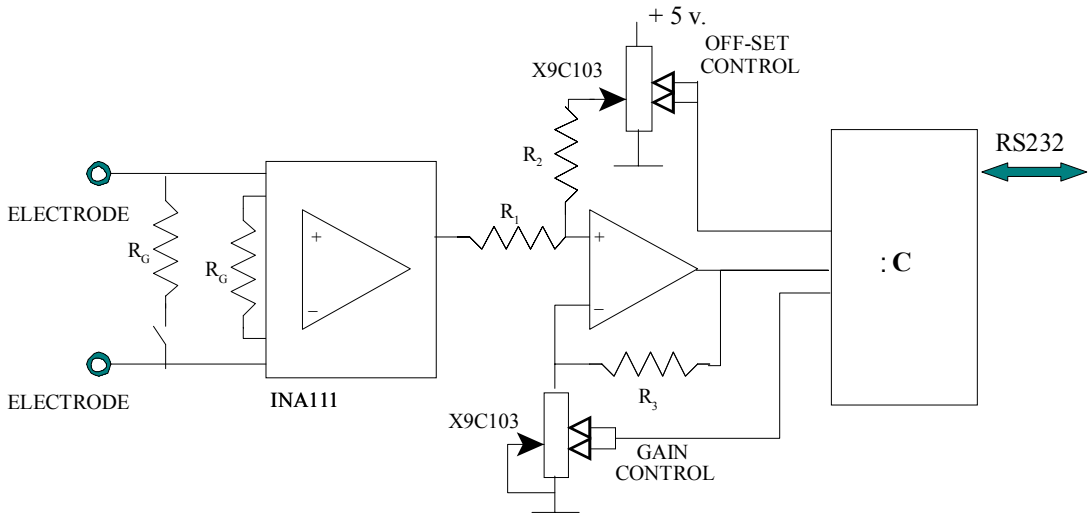

Fig. 3. Adquisition Board

gain adjustments.

The block diagram of the Acquisition Board (Fig. 3) is made up by the following modules:

i.- Microcontroller, controlling the working of the board and communications with the application run on the portable computer. It also digitizes the electrocardiographic signal.

ii. Analogue amplification circuit, based on an instrumentation amplifier. It is a band-pass amplifier between 0.01 Hz and 50 Hz.

iii. Digitally controlled "switching" governed with commands generated by the user of the PC application.

The impedance measuring process is as follows: at the beginning the switch is open, the ECG signal is picked up, digitized and sent on to the portable computer, where a check is made that it has been obtained without artifacts, interference or baseline variations (Fig. 4). N signal cycles (normally N in [10, 20]) are averaged out on the computer, using the following correlation function [7]:

$$\rho = \frac{\sum_{i=1}^{M} x_i . y_i}{\sum_{i=1}^{M} x^2{}_i . \sum_{i=1}^{M} y^2{}_i} \qquad (4)$$

In this way, the new signal-to-noise relation is:

$$SNR_N^2 = N . SNR^2 \qquad (5)$$

The averaged signal (Fig. 5) is digitally processed on the personal computer; the user marks a horizontal line with his/her mouse (reference baseline) and 2 vertical lines, indicating the beginning and end of the averaged cycle. At this moment, the software obtains the E_{OPEN} value. In the next step, the relays is closed and the same process is done, to obtain the $E_{CLOSE,}$ With this information, is possible to apply eq. 3 and obtain the transthoracic impedance.

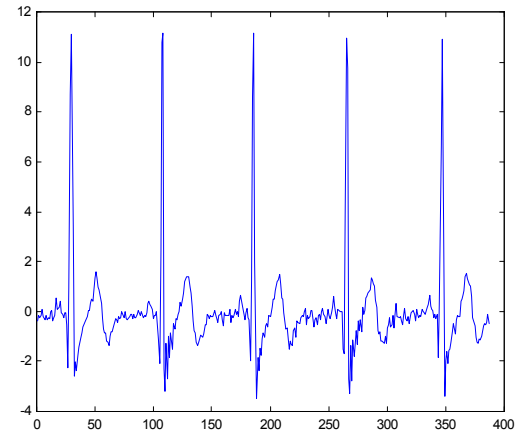

Fig. 4. Digitized ECG signal

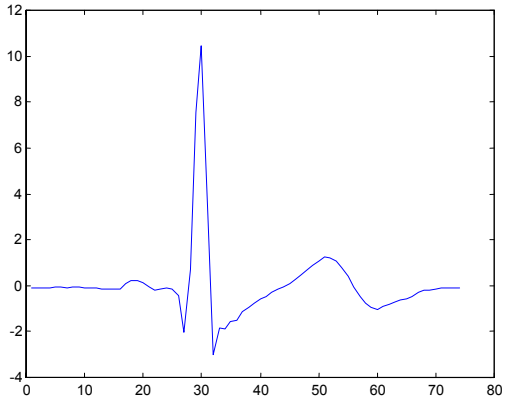

Fig. 5. Averaged ECG signal

In the figure 6 is shown the user interface of the computer.

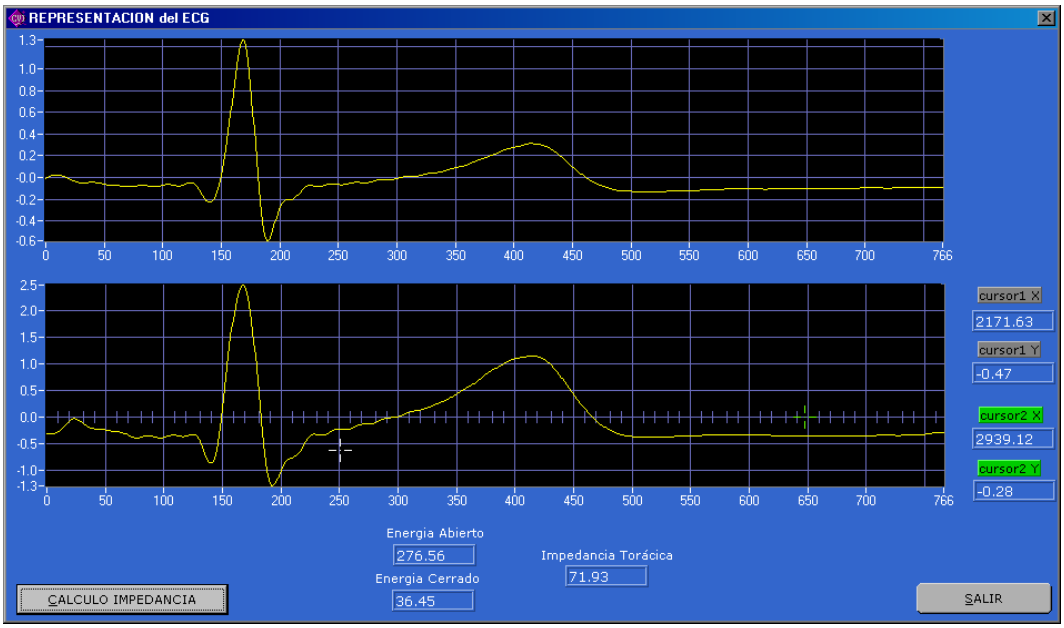

Fig. 6. User Interface

2.1. Population and validation methods

162 patients with high blood pressure were included in the validation study. 66 were male, 10 suffered chronic bronchitis and 15 had suffered thoracic surgery. Informed consent was obtained in each case even though all tests were innocuous for the patients.

For all the patients the same protocol was followed: height and weight were measured. An echocardiographic study was obtained, and immediately several recordings of ECG and thoracic impedance were obtained. Left ventricular mass was calculated using previously validated formulae which use simple measurements of the left ventricle. All measurements were done at least thrice and the mean value was used for statistics. Data were analyzed with the statistical package SPSS 8.0. Correlations between continuous variables were studied by the Pearson´s coefficient. Means were compared by the "t" test for independent variables. A p value of less than 0.05 was considered statistically significant in a two-tailed study.

3. RESULTS

The mean value of thoracic impedance was of 36.1±32.2 Kilo-ohms (rank 1 to 171). Impedance correlated with height (r=0.363, p>0.0001), body surface area (r=0.23, p=0.004), and age (r=-0.16, p=0.046). Greater impedance was observed in male (52.1±40.6 vs 24.8±17.9, p<0.0001), in patients with chronic bronchitis (82.5±51.8 vs 34.0±28.4, p<0.0001) and in patients with previous thoracic surgery (61.8±35.2 vs 34.5±31.2, p=0.01).

The variability of the measurements of impedance for each patient was estimated as the mean of the relative differences between each measurement and the mean value of all. The variability of the measurement for the same individual was of 13.8%±13% of the mean value.

Many different simple and combined measurements of the ECG correlate with left ventricular mass and have been published as valid methods to estimate hypertrophy. Several of these validated ECG measurements were compared with the left ventricular mass obtained with echocardiography. The best correlation between those measurements of the ECG signal and the left ventricular mass was with r = 0.479 (p<0.0001). If the best ECG measurement was corrected by thoracic impedance the correlation with left ventricular mass increased up to r=0.631, p<0.00001. Expressed in clinical terms of sensitivity, for a fixed specificity of 95%, the sensitivity of

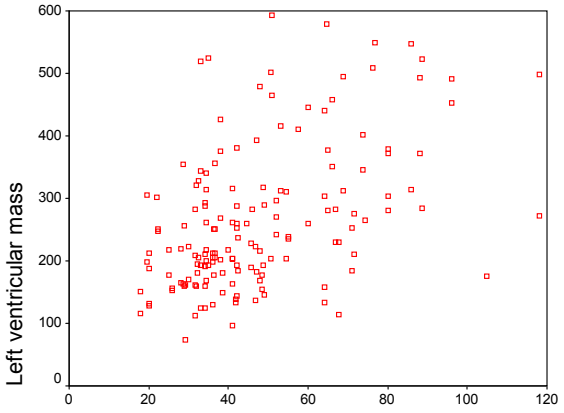

Fig. 7. Maximal area of R wave between V5 and V6

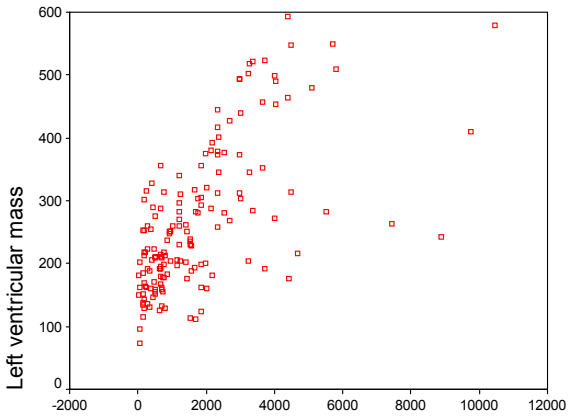

Fig. 8. Maximal area of R (V5/V6) corrected by impedance

the best measurement of ECG was 37% (Fig. 7) and rose to 51% when impedance was used to correct the ECG parameters (Fig. 8). In practical terms, this means that simple ECG measurements can detect only 37% of patients with left ventricular hypertrophy, while the correction of ECG by impedance allowed us to detect up to 51%, in both cases with a 5% of false positive diagnoses.

4. CONCLUSIONS

The main conclusions of this project are:

1.- The system for measuring the transthoracic electrical impedance allows the measurement of a physiological parameter related with anthropometric variables such as height and body surface area, and reflects pathological conditions that theoretically can alter thoracic impedance such as chronic bronchitis and thoracic surgery.

2.- The measurement of transthoracic impedance allows a correction of the ECG derived parameters that increases the correlation with left ventricular mass and improves the senstitivity of ECG for the diagnosis of left ventricular hypertrophy.

5. REFERENCES

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