Measuring Impedance in Congestive Heart Failure

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Abstract. The hospitalization of patients with Heart Failure represents an increasing burden for the healthcare system with more than 23 million worldwide suffering from this disease. In this paper we explore methods to detect fluid retention in the lungs by measuring the thoracic impedance, so that is possible to monitor Heart Failure patients, and physicians can early detect acute episodes. A small and portable device was developed to measure the thoracic impedance of the patient. From the measured thoracic impedance it can estimate the accumulation of fluid in the lungs. This device is a low cost, friendly to use equipment that can be operated by a big range of users: Moreover, it was designed for low power consumption with a rechargeable battery for portable use. The device empower the patient to monitor his own body fluid at home, and a physician can follow him remotely. This procedure would help to drastically reduce the number of hospitalizations and, consequently, improve the quality of life of people diagnosed with Heart Failure.

Keywords. Bioelectrical Impedance; Congestive Heart Failure; Gain Phase Detector.

Introduction

The healthcare system is overloaded with patients hospitalized due to Heart Failure (HF). In the United States of America (USA) more than 5.8 million suffer from HF and over 23 million worldwide. Every year, more than 550,000 cases are diagnosed in the USA [1]. In Portugal there are approximately 260,000 individuals suffering from HF [2]. Elderly are the most affected by HF problems, and considering the worsening risk factor profile for HF in the population (e.g. diabetes, hypertension), aging of the population (due to the longer life span), and the increasing HF prevalence, it is likely that these numbers will increase [3].

HF represents also a considerable burden to the healthcare system being responsible for costs of more than $39 billion annually in the USA. In developed countries, HF hospitalization represents 1-2% of all healthcare expenditures [1].

Measuring body impedance is becoming increasingly available in the clinical setting as a tool for assessing hemodynamics and can be used to measure the thoracic impedance in order to identify patients at risk of acute decompensated heart failure.

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The disease progression is marked by a gradual retention of fluid in the lungs, long before symptoms of disease worsening occur. In order to prevent this significant mortality, morbidity and healthcare expenditures it is necessary to monitor patients at risk, in a way to detect congestion episodes before worsening of symptoms that will lead to hospitalization. There is not available in the market a portable and low cost solution that allows the common user to auto-diagnose in a continuous way, not needing to go to the hospital. This paper proposes a low-cost, low-power and small size solution that can be controlled by a smartphone.

The rest of this paper is organised as follows. Section 1 describes the system modules. In section 2 the experimental results are described followed by section 3 where the system is evaluated. Section 4 draws the main conclusions and points out the future work.

1. System Description

The circuit that generates the excitation source includes a sine wave generator and a Voltage Controlled Current Source (VCCS) and will be responsible for generating the excitation current in a range of frequencies from 20 kHz to 1 MHz. Two equivalent Instrumentation Amplifiers (INA) will be used, one to amplify the signal measured by the pair of voltage electrodes and the other to amplify the voltage drop across the reference resistor, then the INAs two input buffers will be used to match the internal impedance of the Gain Phase Detector (GPD).

The voltage from the pair of electrodes and reference resistor is compared using the GPD which calculates the magnitude ratio and the phase difference. So the low-power feature can be achieved a power circuit is included in the design. Pandlets block will be responsible for tuning the signal generator, acquiring the digital signal from the ADCs and transferring the data to a smartphone using Bluetooth Low Energy (BLE).

![Figure 1. Structural diagram of the device.](image-url)
2. Experimental Results

In order to evaluate the performance of the system, five different RC circuits were prepared, with the hypothesis that the Extracellular Fluid (ECF) and Intracellular Fluid (ICF) act like resistors and the cell membrane like a capacitor in biological tissues [4]; the nominal values of R1, R2 and C are described in Table 1, and the cut-off frequency is calculated for each circuit as proposed by [5] using equation (1).

\[
f_c = \frac{1}{2\pi C} \cdot \sqrt{\frac{R_1 + R_2}{R_1 \cdot R_2}}
\]

Each circuit was measured within a range of 32 frequencies from 20kHz to 1MHz logarithmically spaced (20, 25, 30, 35, 40, 45, 50, 55, 60, 65, 70, 75, 80, 85, 90, 95, 100, 150, 200, 250, 300, 350, 400, 450, 500, 550, 600, 650, 700, 800, 900, 1000 kHz). The frequencies are spaced logarithmically to ensure a proper density of data [6].

Table 1. Parameters for the five measurements.

<table>
<thead>
<tr>
<th>Measurement Number</th>
<th>R1(Ω)</th>
<th>R2(Ω)</th>
<th>C(nF)</th>
<th>f0(kHz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>150</td>
<td>150</td>
<td>10</td>
<td>150</td>
</tr>
<tr>
<td>2</td>
<td>200</td>
<td>200</td>
<td>10</td>
<td>113</td>
</tr>
<tr>
<td>3</td>
<td>200</td>
<td>200</td>
<td>68</td>
<td>16.6</td>
</tr>
<tr>
<td>4</td>
<td>300</td>
<td>300</td>
<td>68</td>
<td>11</td>
</tr>
<tr>
<td>5</td>
<td>270</td>
<td>270</td>
<td>10</td>
<td>83</td>
</tr>
</tbody>
</table>

The values of Phase and Magnitude were stored in a CSV file and the phase-frequency (Figure 3) and magnitude frequency (Figure 4) characteristics were plotted.
It is possible to observe in the phase plot that the characteristic frequency is the point where the phase has its minimal value. In the first stage (low frequencies) the capacitor in the RC circuit works as an open circuit and after the cut-off frequency it starts to work as a short circuit. When comparing the values of the different RC circuits with the values of the measurements it is easy to identify similarities.
However, measurement errors are nearly inevitable, random errors in measurements can be eliminated by performing a set of measurements and calculate the average, but systematic errors could only be eliminated for example by calibration.

3. System Evaluation

During the development, it was decided to develop a PCB with the size of 80 X 60 mm. Most of the chips used were SMD so it could be saved some space. The system has a power consumption of 260 mA when in idle state and it goes to 280mA during measurements. The battery used is a Lithium-ion Polymer rechargeable battery, considering a power consumption of approximately 270mA (alternating between idle and measurement state). The build of materials (BoM) of the measuring device is of 147.11 $ and the price of its PCB is 102.91 $; this price could be decreased in a next version by developing a minor version using passive components of smaller sizes. Our end-to-end solution is completed with the Android App allowing the user to start the measurement and previously receive the values of the measurement.

4. Conclusions

The main objective was to develop a method for measuring the bioelectrical impedance of the thorax so fluid overload in the lungs could be detected long before hospitalization. The use of a BLE communication proved to be very efficient due to its low-cost and low-power consumption, adding the fact that was possible to use the Pandent for this communication. A hybrid solution was designed to allow in the future the integration of other devices resulting in a multi-parameter monitoring system. A user-friendly Android application provides an easy way for the patient to monitor his status. The values of the measurement can be later saved in a database or sent to the caregiver or a physician. The results obtained in the test measurements allowed to conclude the validity of the system, tested in different conditions in a controlled environment. The next step will be to test in a clinical environment with the supervision of a cardiologist. There are some solutions available in the market, but they are expensive, and their big sizes makes them not suitable for a user to have them at home and to continuously monitor himself. During the development of this project, two aspects were always kept in mind: the portability and the low-cost solution. Both of these goals were achieved, having in the end a small size and low power device for less than 300 $ that can be carried anywhere.

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