Design and Evaluation of a Portable Device for the Measurement of Bio-impedance- Cardiograph

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Abstract: Electrical impedance of biological matter is known as electrical bio-impedance or simply as bio-impedance. Bio-impedance devices are of great value for monitoring the pathological and physiological status of biological tissues in clinical and home environments. The technological progress in instrumentation has significantly contributed to the progress that has been observed during the last past decades in impedance spectroscopy and electrical impedance cardiograph. Although bio-impedance is not a physiological parameter, the method enables tissue characterization and functional monitoring and can contribute to the monitoring of the health status of a person. In this paper an inexpensive portable multi frequency impedance cardiograph device based on impedance spectroscopy technique has been developed. By means of this system the basic thoracic impedance range and the heart-action-caused changes of impedance can be measured and the hemodynamic parameters of the heart function can be determined. This system has small size and low current consumption. The impedance cardiograph signals of the electrodes configuration by Sramek, Penney and Qu in this work was measured; compared and summarized. The differences of the measuring method, the schematic circuit diagram, the measurement results and area of application between impedance cardiograph and impedance spectroscopy were discussed and compared. The performance of this sensor-system was evaluated.

Keywords: impedance cardiograph (ICG); impedance spectroscopy (IS); electrocardiography (ECG); electrical bio-impedance of the chest; cardiac outputs; hemodynamic; non-invasive diagnostic

1. Introduction

Evaluation of the hemodynamic parameters of a patient has always been a subject of interest to clinicians. It has been difficult to capture hemodynamic parameters and invasive methods have been used to measure them. These invasive techniques are expensive, time-consuming, demand complicated equipment and trained staff and are not always possible to use because of the condition of the patient, which maybe either too serve or else too good to run the risks associated with invasive techniques[1].Therefore they are not suitable for long time or repeated measurements because of their invasive nature. Several non-invasive techniques capable of monitoring cardiac activity were in past two decades developed. They are ultrasound Doppler, magnetic resonance imaging, and impedance cardiograph (ICG). However, all these techniques except ICG are not suitable for long-term continuous monitoring of cardiac activity. Impedance cardiograph is a simple, inexpensive, and non-invasive method for hemodynamic parameters measurements.

The method is based on changes in the electrical resistance of the chest during heartbeat [2]. Previous Studies suggest that various physiological variables contribute to the impedance changes. The most commonly mentioned contributors are: Enlargement of the volume of aorta, enlargement of the volume of the blood in the pulmonary circulation and laminar blood flow in the large vessels [3]. It is generally accepted that ICG originates from a combination of blood volume change, rearrangement of current conductors, redistribution of current density, and resistivity

change of flowing blood within the measuring object. ICG was developed into clinical practice by Kubicek and Colleagues using their proposed equation and band electrode array [Fig.1.A]. However, the band electrodes are not practical for use. The two electrodes around the neck can produce an annoying, or even a choking sensation. This maybe cause increased apprehension in some patients; which could be harmful in patients with cardiac disease. Furthermore, it is difficult to place the full band electrodes on patients with chest burns or on patients recovering from cardiac surgery who have incisions and dressings or tubes and lines where the electrodes need to be applied [5]. New electrode arrays were introduced using disposable spot electrodes. In modern ICG systems electrode configurations by Sramek, Penney and Qu are used [6] [7] [8]. In electrode configuration by Sramek four electrodes are used to deliver the electrical current, which are known as "current electrodes". Another four electrodes are used to measure the voltage changes, which are known as "voltage electrodes". The electrodes are positioned symmetrically on both sides of the patient's neck and the chest [Fig.1.B]. An alternating current flows along the current electrodes through the measured body segment and the voltage electrodes measure the voltage changes. For the electrode configuration by Penney the electrode array, as illustrated in Fig.1.C, uses for ECG spot electrodes. Two are placed at the base of the neck, separated by 6 cm and centered about the prominence of C-7. Two more electrodes are placed below the heart on the left anterolateral chest surface. One is placed at the end of the ninth intercostals space, near the mid-clavicular line. The other is 8 cm far away from the first, in the tenth intercostal space, near the mid-axillary line. The current is passed between the electrode on the right of the neck and that at the end of the ninth intercostals space. The voltage difference is measured between the remaining electrode pair [7]. Qu et al. developed a new plot electrode configuration. One voltage electrode is placed 4 cm above the clavicle on the anterior surface of the neck; a second voltage electrode placed over the sternum at the lever of the fourth rib; a current electrode placed at the lever of the fourth vertebrae on the posterior surface of the neck; a second current electrode placed at the lever of the ninth thoracic vertebrae on the back. This plot electrode configuration yielded improved signal quality because the placement of sensors along the midline minimized movement and breath artifacts during exercise [9] [10]. The impedance cardiograph signals of the electrodes configuration by Sramek, Penney and Qu in this work was measured; compared and summarized.

2. Methods

The hardware module in this work is based on the tradition method where a constant current I(t)(approximately 1mA at 100k Hz) is fed into current electrodes (see Fig. 2) and the resulting voltage U(t) is measured through inner voltage electrodes, thereby allowing the calculation of the electrical bio-impedance Z(t):

$$Z(t) = U(t)/I(t) \qquad\qquad (1)$$

To realize this simple method, the following issues have to be kept in mind: the current fed into the patient needs to be a well defined constant signal with fixed magnitude, time and frequency. A direct digital synthesis (DDS) IC is used to generate the sine wave signal. This sin wave voltage can be fed into the patient through the ECG Electrodes. The resulting patient voltage is detected by an instrumentation amplifier and the current is measured by the other instrumentation amplifier at the same time. Then the gain phase detector can be used to detect the magnitude and phase of the bio-impedance of the selected segments. The magnitude of bio-impedance consists of two important signals: the base thoracic impedance and thoracic impedance changes. There are some noises in this signal: the impedance changes because of breath and motion

which should be eliminated. Then a differentiator is used to derivate the thoracic impedance changes. All of the analog signals are sent to analog to digital convert of the microprocessor MSP430f1611 and over there digital signal processing for calculation of hemodynamic parameters can be programmed. MSP430f1611 is the microcontroller configurations with two build-in 16-bit timers, the fast 12-bit analog to digital converters, dual 12-bit digital to analog converters, universal serial synchronous/asynchronous communication interfaces (USART), I2C, DMA and 48 I/O pins.

This system can be considered as amplitude modulation in communication system for the information transmission (Fig. 9). The stimulated signal is considered as carrier wave. The measured impedance signals can be considered as the original signals. Therefore the output signals of this system can be considered as the modulated signal. The output signals can be separated into the different parts with the gain phase detector and base line system. The measured impedance consists of three different parts: the base thoracic impedance Z_0, heart action caused impedance changes ΔZ_{heart} and breath caused impedance changes ΔZ_{breath}, the base line system can be used to separate these three different parts from the measured signal. Like we know that the heart beat frequency f_1 is from 1 to 2.5 Hz and the breath frequency f_2 is from 0.25 to 0.43 Hz. In equation (1) the H is the amplitude of heart action caused impedance changes ΔZ_{heart} and B is breath caused impedance changes ΔZ_{breath}. The noise comes from the system and stimulus signal.

$$Z = Z_0 + \Delta Z_{heart} + \Delta Z_{breath} = Z_0 + H \cdot \sin(2\pi \cdot f_1 \cdot t + \varphi_1) + B \cdot \sin(2\pi \cdot f_2 \cdot t + \varphi_2) + noise \quad (1)$$

An automatically balancing circuit is used to detect the thoracic impedance which consists of two low pass filters with the cut off frequency 22Hz and 0.7Hz and a differential amplifier (Fig. 10). With the low pass filter with cut off frequency 22 Hz the noise signals can be eliminated. With another low pass filter with cut off frequency 0.7 Hz the heart action caused impedance changes is ejected and we can get the base thoracic impedance and breath caused impedance changes. Then with the digital signal processing these two signals can be separated and saved. A differential amplifier is used to acquire the heart action caused impedance changes. Therefore these three different parts can be separated.

3. Results and Analysis

In this capital the electrode configurations by Sramek, Penney and Qu are measured and compared with this ICG System. As illustrated in Fig. 3 the base thoracic impedance Z0 is 28 Ω (2.8V, G = 100). The thoracic impedance changes is minimal 0.5 Ω and maximal 0.8 Ω which is ca. 3% of the base thoracic impedance (0.6 V to 0.96 V, G = 1200). The four current spot electrodes are symmetrically placed on both side of the neck and chest. Therefore that generates homogeneous electrical field and the impedance of electrode can be neglected so that these can improve the accuracy of measurement. The reproducibility of measurement of this electrode configuration is best than the other plot electrode configurations. The base thoracic impedance Z0 is 18 Ω (1.8 V, G = 100) (Fig. 4). The thoracic impedance changes is minimal 0.5 Ω and maximal 0.8 Ω (0.6 to 0.96 V, G = 12000) and is ca. 4% of the base thoracic impedance. The base thoracic impedance by this configuration is smaller than by Sramek. As showed in Fig. 5 the base thoracic impedance Z0 is 18 Ω as same as the electrode configuration by Penney (1.8 V, G = 100). The thoracic impedance changes is minimal 0.6 Ω and maximal 0.9 Ω (0.72 to 1.1 V, G = 1200) and ca. 5% of the base thoracic impedance which is greater than another electrode configuration. In this work

the influence of breath for the thoracic changes by these three spot electrode configurations is evaluated. As illustrated in Fig. 6-8 we can make a conclusion, that Qu spot electrode configuration has the best signal noise ratio than another two configurations. Because from Fig. 10 and 11 we can see that they have the same constant basis line.

4. Conclusions

The electrodes configurations and signal processing play an important role in signal reproducibility. In this study the results by different electrodes configuration (Sramek, Penney and Qu) are measured and compared with the same circuit. The results of electrodes configuration by Sramek are similar as band electrodes configuration. But this electrodes configuration is sensitive to breath and motion. The results of electrodes configuration by Qu have best accuracy and the breath and motion have no effect on the results of this electrodes configuration. In this work a basis line system is used to eliminate the effect of breath and motion on the results. With this system the section of heart-action caused changes of impedance and of breath caused changes of impedance can be separated. That means, the noise of breath caused changes of impedance not only can be eliminated from the basic thorax impedance and heart-action caused changes of impedance but this signal can be also measured in this system. In the future, more advanced techniques such as adaptive filter, specific model and digital signal processing for calculation of hemodynamic parameter can be implemented for detection to increase the accuracy and reliability. The SD card can be used in the future to storage the large data. A fewer components will be hoped to use and therefore as few as possible power can be used that a long time monitoring system can be achieved in the future.

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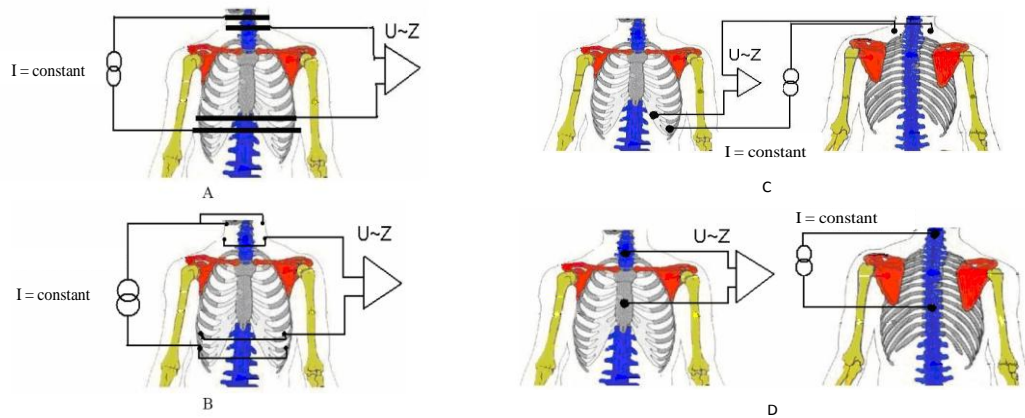
Figure 1. A: band electrode configurations by Kubicek; B: 8 spot electrode configurations by Sramek; C: spot electrode configurations by Penney; D: spot electrode configuration by Qu
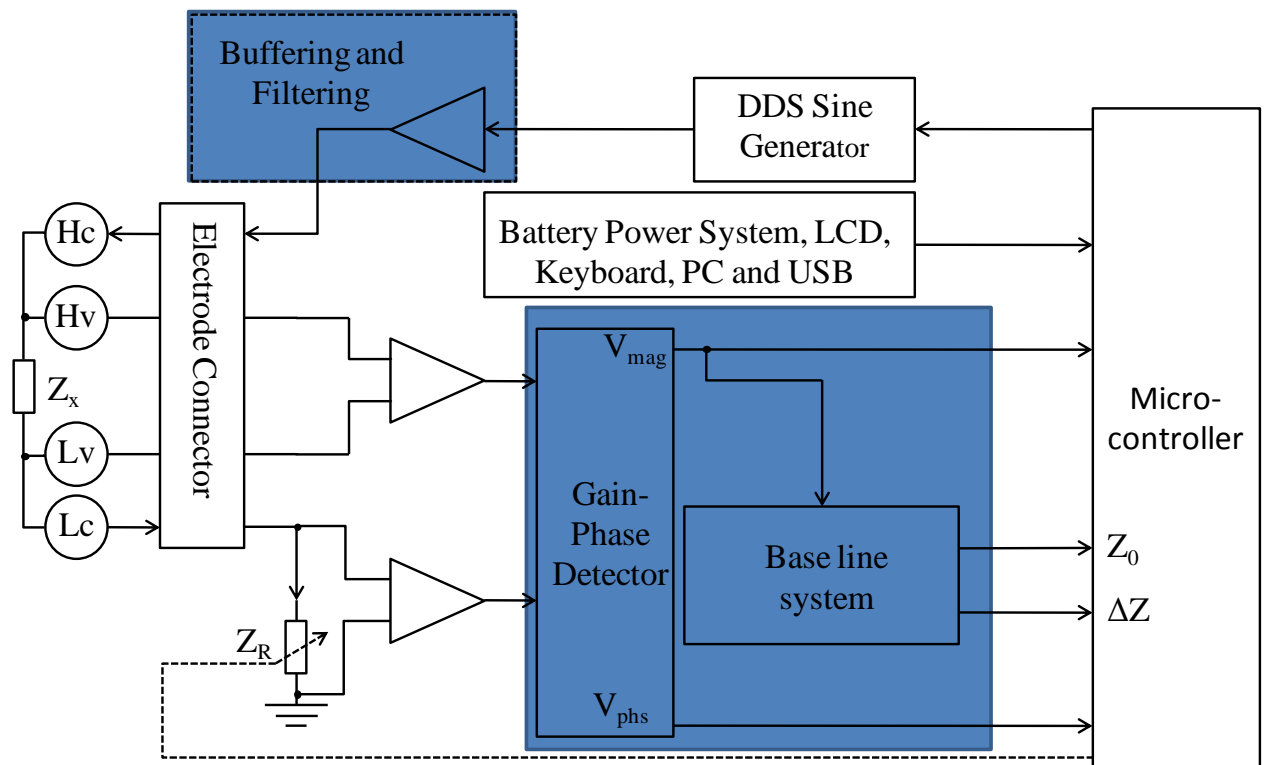

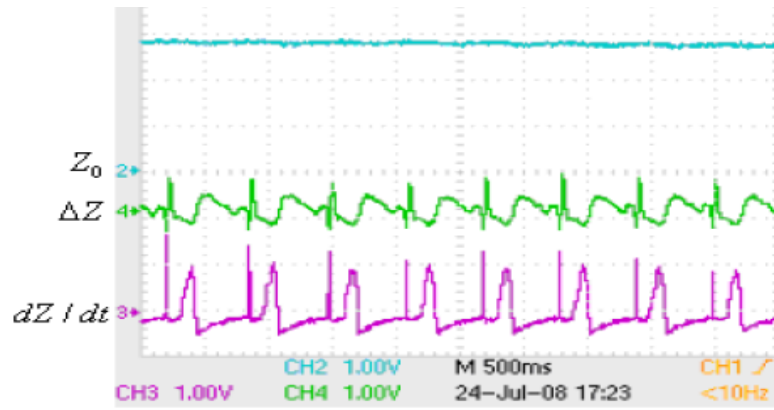Figure 2. The block diagram of ICG System

Figure 3. Results of Sramek electrode configuration
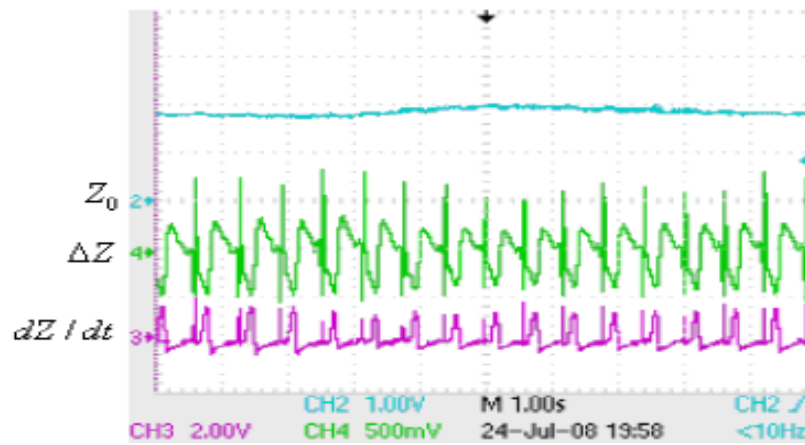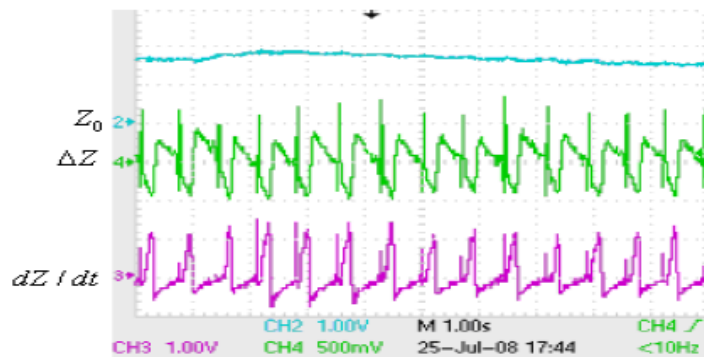

Figure 4. Results of Penney electrode configuration

Figure 5. Results of Qu electrode configuration

Figure 6. Results of Sramek electrode configuration by
a) normal breath   b) tief breath

Figure 7. Results of Penney electrode configuration by
a) normal breath   b) tief breath

Figure 8. Results of Penney electrode configuration by
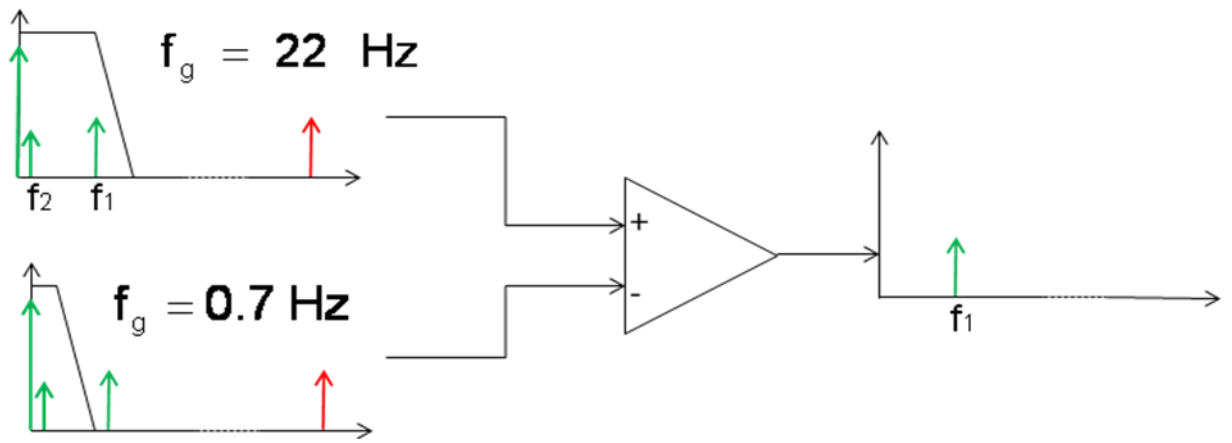a) normal breath   b) tief breath

Figure 9. Structure diagram of a Bio-impedance measurement system
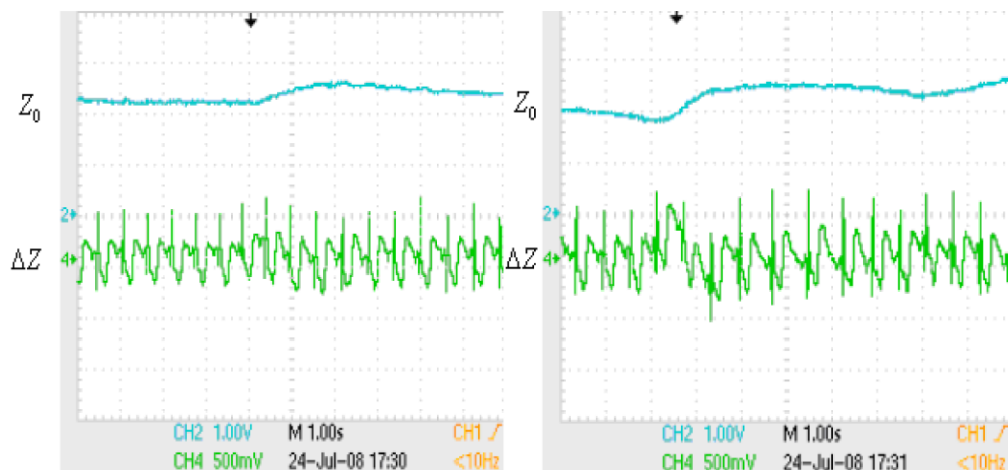
Figure 10. Autobalancing circuit