Design of Bioimpedance Monitor and Its Application to Atrioventricular Delay Optimization

V Vondra, I Viscor, J Halamek, P Jurak

Institute of Scientific Instruments of the ASCR, vvi, Brno, Czech Republic

Abstract

We introduce our conception of a two-channel bioimpedance monitor for the impedance cardiography. We are describing the design of a homemade device which is based on a direct digital synthesizer and a digital down converter. This solution enables us to obtain a high-quality bioimpedance signal, which can be used for computing the cardiac output beat per beat. Our device allows simultaneous measuring of the complex impedance in two arbitrary directions on a human/animal body without any interference between the channels. We tested our device during a study of measuring the cardiac output at patients with implanted pacemakers. The cardiac output is measured with respect to the atrio-ventricular delay. We have scored the maximum relative cardiac output for each patient. The result of the study of 26 patients is presented in this paper.

1. Introduction

The bioimpedance measurement for analysing the cardiac system was introduced by Kubicek more than 40 years ago [1]. The main advantage is the noninvasive process of this measurement.

The main principle of the bioimpedance measurement is based on applying a weak electrical current to patient´s body and scanning the induced electrical voltage. Bioimpedance techniques in medicine have many applications from skin resistance measurement through cardiac system analysis to impedance tomography [2].

The impedance cardiography (ICG) measures impedance changes in the thorax to calculate the stroke volume or the cardiac output respectively as the final quantity. The evaluation of the stroke volume from the bioimpedance signal has gone through its evolution from the former Kubicek equation, through Sramek equation and finally to Bernstein equation [3], and it seems to go on. Lot of studies have shown us that ICG has similar accuracy and reproducibility like other methods for the cardiac output evaluation, e.g. the invasive and expensive thermodilution, echocardiography requiring a high skilled and experienced operator [4], [5]. The essential attribute of ICG for our purpose is its mentioned noninvasive type. We concentrate on analyzing the cardiovascular system properties and developing noninvasive diagnosis methods [6]. For this purpose we do a complex measurement, and the stroke volume computed from ICG is one of the most important parameters which are used together with other biological signals (ECG, breathing, blood pressure, phonocardiogram, etc.) to evaluate the condition of the cardiovascular system. Because of the experimental and research type of our measurement and thus the necessity of setting all parameters of the measurement arbitrary, we decided to build a homemade two-channel bioimpedance monitor which enables us to have all the measurement parameters and further signal evaluation (continuous stroke volume and/or cardiac output beat per beat) fully under our control.

For testing and evaluation of our device we made a pilot study oriented on an atrio-ventricular delay (AVD) optimization. The atrio-ventricular delay is one of the important parameters of heart pacing. It has a direct influence on the cardiac output. A suboptimal AVD setting could lead to a significant decrease of the cardiac output [7], [8], [9]. The basic prerequisite for setting of an optimal AVD is the detailed knowledge of its influence on the cardiac output. The study is concerned with the measurement of the cardiac output with respect to AVD at patients with implanted pacemakers. The result should lead to a better pacemaker setting with CRT patients who are treated at I. Internal Cardioangiologic Clinic, St. Anna's University Hospital in Brno. The noninvasive ICG based method is also the most comfortable for patients. This could also increase the measurement accuracy because the stress for the patients is minimized. With the proposed device we have measured the cardiac output by 26 patients and the result of this study is presented here.

2. Methods

The thorax electrical impedance is measured with the standard four-pole method which is less sensitive to the

electrical property of the electrodes and their coupling with the skin. In the first channel we use the current source for the outer two electrodes on patient's neck and the groin. The voltage is monitored by the inner electrode pair placed in the neighbourhood of the clavicle and under the rib cage. All electrodes are placed on the left side of the thorax. The second channel is applied perpendicularly – current electrodes over the shoulders and voltage is monitored on the chest – all four electrodes are placed in one horizontal line. The aim is to monitor the cardiac output into the aorta and the pulmonary artery [10] simultaneously. In general each channel can be placed on an arbitrary position on the body. E.g. to study the arterial system elasticity, one channel can be placed on the thorax and the second one on an upper or lower limb to evaluate the shape and time relations of the blood flow in the aorta and in the limb (pulse wave).

Measuring of more than one channel brings up the problem of their separation. We have chosen the so called frequency multiplex which means that each channel i has its own carrier frequency f_{ci}. The frequencies of all the channels should be as close to each other as possible to minimize the frequency dependence of all the channels. The other requirement of the bioimpedance system is a high dynamic range of the whole device because of very small changes of the bioimpedance induced by the heart activity with respect to the overall static bioimpedance of the body and the breathing influence impedance contribution.

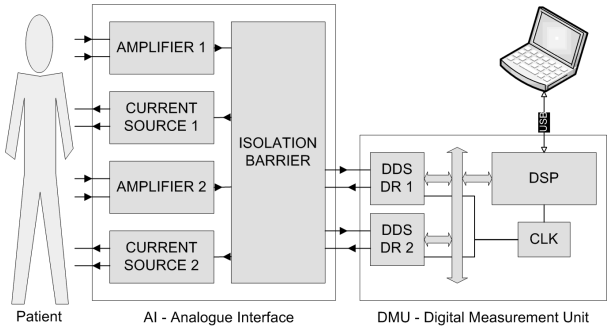

Fig. 1. Block diagram of the two-channel bioimpedance monitor.

The main idea is to use of a couple of digital transmitters and receivers which are controlled by a common frequency standard. Then a coherent signal demodulation is enabled. This solution is able to fulfill both demands: the narrow band signal processing and the high dynamic range. The block diagram of the proposed device is shown in Fig 1. The bioimpedance monitor consists of three main parts: a digital measurement unit (DMU), an analogue interface (AI) and a standard personal computer (PC). The whole system is controlled by a special software in the PC. DMU generates the measurement carrier frequencies for both channels. Both DR and DDS are controlled by the digital signal processor (DSP) which also communicates with the hosting PC. The CLK block is the source of a common clock signal for all digital parts of the DMU. The exact digital control of DDS and DS parts together with CLK guarantees the phase coherence of the whole device. An analogue interface is used to connect the DMU input and output signals to the patient.

The crucial part of DMU is the DDS+DR block. This consists of a direct digital synthesizer (DDR) and a digital receiver (DR). This solution was chosen for its frequency accuracy and stability. The output carrier frequency is generated digitally in the digital synthesizer of the output part and an analogue signal is obtained in the digital to analogue converter (DAC). Digitalizing of the measured bioimpedance signal is done by an analogue to digital converter (ADC) followed by a digital down converter – the input part (DR). DR parts use the high oversampling principle and this has a major influence on the high signal to noise ratio of the ICG signal and good separation of both channels. We achieved these DMU parameters: DDS – output frequency 0.01-120 MHz, narrowband SFDR (5 kHz) 86 dB; DR – input frequency bandwidth 0.01-130 MHz, dynamic range 148 dBFS/Hz, jitter lower than 0.5 ps [8]. The carrier frequency resolution for both DDS and DR is 0.01 Hz. The analogue interface limits the bandwidth up to 120 kHz. The output sampling frequency can be set up to 10 kHz (we use 500 Hz for most applications). Both parameters are reasonable for the ICG purpose. The measurement current can be set up to 5 mA RMS.

A PC-type computer obtains - from the DMU via USB - digital data representing the complex valued voltage on the sensing electrodes. The magnitude and the phase or the real and imaginary parts of the bioimpedance signal are calculated by means of Ohm's law. From this bioimpedance signal the desired quantity is calculated.

All 26 patients included in the pilot study have severe heart failures. They all have implanted pacemakers working in the biventricular mode. All participants underwent a pacemaker device implantation one or more years before this study's measurement and they are in a stabilized state. The patients were investigated during their regular annual inspection in the clinic.

The ICG based cardiac output measurement protocol was organized as follows: The patient lied on an examination bed in the supine position. He/she had been instructed to be at rest with natural breathing during the whole time of the measurement. The following biological

quantities were recorded continuously: ECG, blood pressure, phonocardiogram (PCG) and bioimpedance of the thorax. The CRT device was in the DDD mode. The heart rate was set to 80 beats per minute. CO was measured for the following AVD intervals: 75, 100, 120, 140, 160, 180, 200, 225, and 250 ms. Measuring all the AVDs was performed in one series. The order of the AVD intervals was randomized for each patient separately. The ECG was measured by a standard 12 lead ECG device, the blood pressure was measured continuously by the Ohmeda continuous blood pressure monitor from a digit on the patient's right hand. PCG was monitored by a home made PCG monitor with a transducer placed between the 2nd and 3rd rib on the right part of the patient's thorax.

All biological signals were digitized and recorded to the personal computer and processed off-line subsequently after the measurement. For the evaluation of the stroke volume (SV) we applied the commonly used Kubicek equation [3]. The left ventricular ejection time was derived from the PCG signal by filtering and subsequent detecting the S1 and S2 heart sounds [5], [11]. The cardiac output (CO) was then calculated with the equation CO=SV*HR. The heart rate (HR) was derived from the ECG signal for higher accuracy. The final CO for the particular AVD interval was the mean value of CO over the whole interval of measuring or for a part of that interval in which all signals (ECG, bioimpedance, PCG) could be processed. For the evaluation of CO we have developed semi automated routines in MATLAB.

Each AVD interval was evaluated separately and AVD for maximal CO was scored for each subject. For the purpose of comparing the results among the patients the CO was normalized to the maximal CO for each patient.

3. Results

The device was built and tested successfully on the group of 26 patients aged 68.3±9.9 years. There were 13 female and 13 male patients. The result average AVD for the maximal CO of the whole group was 156±52 ms for the DDD pacemaker mode. All patients were paced at HR=80 beats per minute during the measurement. The distribution of the maximal CO versus the AVD interval is shown in Fig. 2. From the distribution it is obvious, that there is no common optimal AVD for all of the patients or the majority of them. The maximum value of CO is an individual property. During the evaluation of CO we found several typical curves of the CO versus AVD. Fig 3. shows the measured data of patient No. 6. These values are typical for a group of patients who are very sensitive to proper setting of AVD. In this case an improper AVD setting could decrease CO to 50 % from its maximal value. Another relatively big group of patients has a similar CO versus AVD setting as shown in Fig. 4, with data from the patient No. 15.

This representative case demonstrates that CO is not sensitive to AVD setting in a relatively large range. Changing the AVD from 100 ms to 250 ms caused changes in CO less than 15 % from the maximum value.

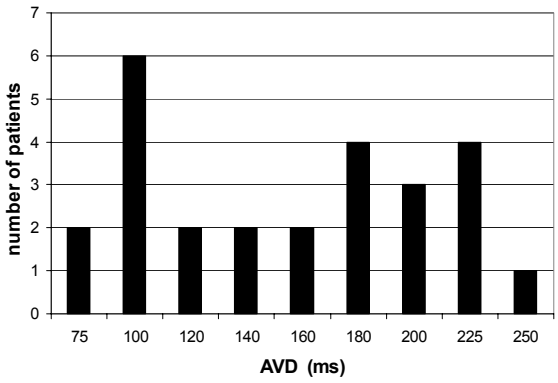

Fig. 2. Distribution of maximal cardiac output.

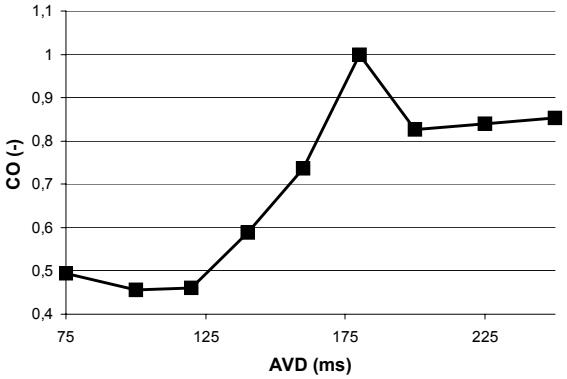

Fig. 3. Cardiac output versus AVD for patient No.6. Improper setting of AVD could cause drop in CO more than 50 %

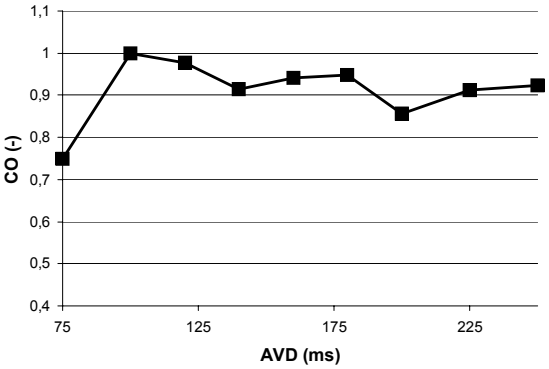

Fig. 4. Cardiac output versus AVD for patient No.15. Setting AVD in large interval from 100 ms to 250 ms causes changes in CO less than 15 %.

The last figure, Fig. 5 represents only one case in which the measurement of CO leads to an unpredictable shape of the CO curve. Instead of a maximum, patient No. 12 has the minimum in the centre. CO raises with either shortening or prolonging the AVD and the maximum CO is then for AVD = 250 ms.

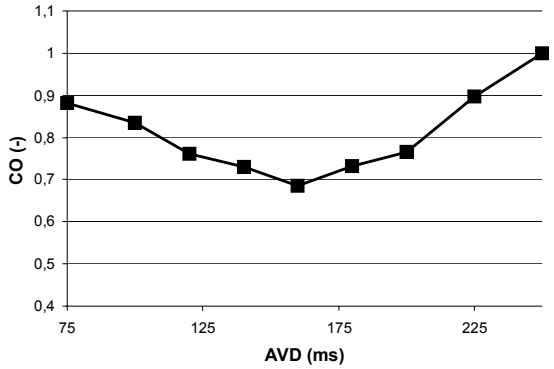

Fig. 5. Cardiac output versus AVD for patient No.12. Significant minimum for AVD = 160 ms. Co increased by either shortening or prolonging of AVD.

4. Discussion and conclusions

The two channel bioimpedance monitor for the impedance cardiography was designed, made and tested successfully. The device is used for research purposes. The complex output two-channel signals enable the beat per beat analysis in two independent channels simultaneously.

The pilot study of a CO measurement at 26 patients with implanted pacemakers in the DDD mode is presented. The average AVD for maximal CO of the whole group was 156±52 ms, but from the distribution of the maximal CO versus AVD it is evident that this parameter is very individual. Setting of the AVD in a large interval does not have to be essential with a part of the group while another part of the group of patients is very sensitive to the proper AVD setting.

The limitation of this study arises from the limited number of patients. We still continue with this study. The results of this study are intended to optimize the AVD setting by means of the ICG method in clinical practice.

Acknowledgements

This work was supported by the Grant Agency of the Academy of Sciences of the Czech Republic under Grant GAAV CRIAA200650801 and by the Grant Agency of the Czech Republic under Grant 102/08/1129.

References

[1] Kubicek WG et al. Development and evaluation of an impedance cardiac output system. Aerospd. Med. 1966; 37: 1208-1212.

[2] Bourne JR (ed.). Bioelectrical Impedance Techniques in Medicine. Critical Reviews in Biomedical Engineering 1996; 24: 223-678.

[3] Bernstein DP, Lemmens HJM. Stroke volume equation for impedance cardiography. Medical & Biological Engineering & Computing 2005; 43: 443-450.

[4] Schmidt C et al. Comparison of electrical velocimetry and transoesophageal Doppler echocardiography for measuring stroke volume and cardiac output. British Journal of Anaesthesia 2005; 95:603-610.

[5] Woltjer HH et al. Standardization of non-invasiveimpedance cardiography for assessmentofstroke volume: comparison with thermodilution. British Journal of Anaesthesia 1996; 77: 748-752.

[6] Halamek J et al. Variability of Phase Shift Between Blood Pressure and Heart Rate Fluctuations. Circulation 2003; 108: 292-297.

[7] Sawhney NS, Waggoner AD, Garhwal S, Chawla MK, Osborn J, Faddis MN. Randomized prospective trial of atrioventricular delay programming for cardiac resynchronization therapy. Heart Rhythm 2004; 1: 562–567.

[8] Kerlan JE, Sawhney NS, Waggoner AD, Chawla MK, Garhwal S, Osborn JL, Faddis MN. Prospective comparison of echocardiographic atrioventricular delay optimization methods for cardiac resynchronization therapy. Heart Rhythm 2006; 3:148-154.

[9] Melzer C, Knebel F, Ismer B, Bondke H, Nienaber CA, Baumann G, Borges AC. Influence of the atrio-ventricular delay optimization on the intra left ventricular delay in cardiac resynchronization therapy. Cardiovasc Ultrasound. [Online] 2006; 4.

[10] Nowakonwski A, Palko T, Wtorek J. Advances in electrical impedance methods in medical diagnostics. Bulletin of the Polish academy of sciences Technical Sciences 2005; 53: 231-243.

[11] Kumar D, Carvalho P, Antunes M, Henriques J, Eugenio L, Schmidt R, Habetha J. Detection of S1 and S2 Heart Sounds by High Frequency Signatures. Anual International Conference of the IEEE EMBS New York 2006; 28: 1410-1416.

Address for correspondence:

Vlastimil Vondra
Institute of Scientific Instruments of the ASCR, v.v.i.
Kralovopolska 147, 612 64 Brno, Czech Republic
vond@isibrno.cz