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DEVELOPMENT OF AN EXPERIMENTAL COMPLEX FOR VERIFICATION AND CALIBRATION OF MEASURING TOOLS OF HEMODYNAMICS PARAMETERS BASED ON BIOIMPEDANCE CHANGES
РОЗРОБКА ЕКСПЕРИМЕНТАЛЬНОГО КОМПЛЕКСУ ПЕРЕВІРКИ ТА КАЛІБРУВАННЯ ЗАСОБІВ ВИМІРЮВАННЯ ПАРАМЕТРІВ ЦЕНТРАЛЬНОЇ ГЕМОДИНАМІКИ НА ОСНОВІ БІОІМПЕДАНСУ

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Abstract*. An article describes research of conditions and development of an experimental complex for check and calibration of measuring devices of parameters of the central hemodynamics on the basis of bioimpedance changing. The most promising and widely used method is the method of impedance cardiography, which consists in passing a low-current (0.5… 5 mA) high-frequency (20-200 kHz) signal through the thorax, and measuring the voltage drop due to dynamic changes in the bioimpedance of the chest under time of ejection of blood by the heart. The possibility of modeling the impedance of the thorax was investigated, and on the basis of the obtained results, an electrical equivalent circuit was developed. The hardware and software parts of the experimental complex has been developed, which allows to model the basic impedance of the thorax and its dynamic change during the ejection of blood by the heart in the range from 50Ω to 1000Ω. The complex also includes an ECG signal generator, as most measuring devices of this method use the ECG signal curve to find reference points for further calculations.*

Key words: *hemodynamics, cardiac output, parameter monitoring, thoracal electrical bioimpedance.*

INTRODUCTION

Determining the parameters of hemodynamics is an important and acute topic nowadays, because only data of these parameters can accurately characterize the state and efficiency of the human cardiovascular system. The main parameter is the minute volume of blood circulation, or cardiac output (CO) — amount of blood that the heart pumps in 1 minute.

Historically, methods for determining hemodynamic parameters were invasive, which required the introduction of a measuring instrument into the circulatory system. Invasive methods have many disadvantages, such as the possibility of infection, the possibility of damage to vital organs, the dependence of the result on the actions of the operator, and the impossibility of long-term measurement. But to date, the method of thermodilution, which is invasive, is used, and recognized as the "gold standard" for measuring the parameters of human hemodynamics. Today, non-invasive measurement methods are being developed that will obtain data from electrodes attached to the human body without damaging it. These methods eliminate the disadvantages of invasive, but still not recognized their correlation with the method of thermodilution.

When developing new devices of measuring cardiac output based on changes in bioimpedance, there is a need to perform comprehensive testing and calibration. Since the primary signal is electrical resistance, it is possible to use an electrical equivalent circuit to build a model of chest impedance.

Not many scientific papers describe this topic. Proposing a new approach to the experimental complex, existing publications and developments were studied [1,2,3,4].

BIOIMPEDANCE MEASUREMENTS

The most promising non-invasive method of measuring cardiac output is the method of impedance cardiography, which is based on measuring changes in bioimpedance. During systole, blood enters the chest area, which causes a change in bioimpedance. The obtained values of the change in impedance can be used to estimate the amount of cardiac output using appropriate model of blood flow.

The method of impedance cardiography uses two pairs of electrodes, one of which is conductive and the other measuring. Low-amplitude (0.5… 5 mA) high-frequency (20-200 kHz) current is supplied through electrodes, as at frequencies of 20… 200 kHz blood is the most electrically conductive tissue structure. The voltage is measured at the other pair of electrodes. The obtained data make it possible to plot the impedance change. Using the graph of impedance changing, the time derivative of the change, and certain characteristic points, it is possible to calculate the magnitude of stroke volume and cardiac output, as shown in Figure 1.
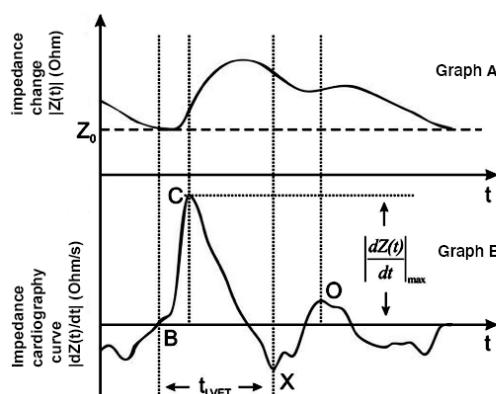

Figure 1. Graph of thoracic impedance and ICG wave with their characteristic points

Characteristic points for determining the cardiac output: point C is the maximum of the curve, point B is the intersection of the curve with zero value, and point X is the local minimum after point C [5]. Several mathematical models have been proposed by various authors to estimate the impact volume using these characteristic points. The most common is the Bernstein-Sramek equation (1):

$$SV = \delta \cdot \frac{(0.17 \cdot H)^3}{4.2} \cdot \left| \frac{dZ}{dt} \right|_{max} \cdot \frac{t_{LVET}}{Z_0} . \tag{1}$$

In this equation, SV is the stroke volume, ie the amount of blood per heart contraction, the distance between the electrodes H is assumed to be 17% of the patient's height. The coefficient δ is the actual weight of the patient divided by the ideal weight, which was derived by Bernstein [6], T_{Lvet} - time of ejection of blood in

the left ventricle, the interval B-X in Figure 1, Z_0 - base chest impedance, $(dz/dt)_{max}$ - maximum derivative in time [7].

$$CO = SV \cdot HR. \tag{2}$$

To determine cardiac output (CO), the stroke volume (SV) must be multiplied by the heart rate (HR).

ELECTRODES

The patient's electrodes are used to conduct electronic charges between the measuring device and the human body. In clinical practice are mostly used electrodes with a gel. For further modeling, it is necessary to present the patient-electrode contact in the form of an electrical equivalent circuit. In the general case, the circuit consists of three main components: the conductivity of human tissues and fluids, the conductivity of human skin and metal part of the electrode in contact with the skin [8]. An additional circuit is the electrolyte between the metal part of the electrode and the skin: a solid, viscous or liquid gel. The equivalent scheme (Fig. 2) presents by RC circuit describing the electrode, the resistance of the electrolyte between the conductive metal part of the electrode and the layer of the epidermis, and other circuits that are result of the tissues structure.
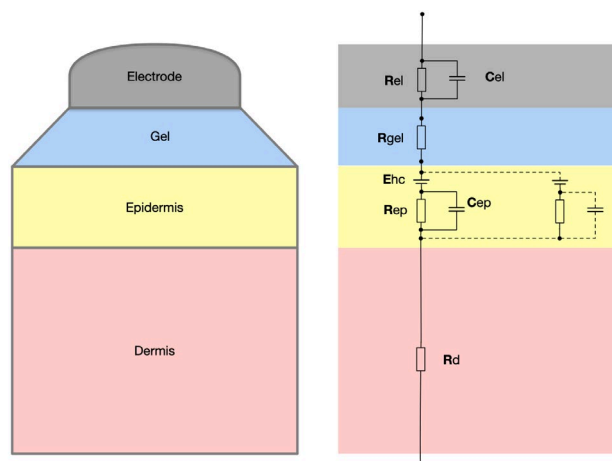

Figure 2. Equivalent scheme of electrode-patient interface

The total resistance for entire equivalent patient-electrode circuit can be calculated by the formula:

$$Z_{el} = \frac{R_{el}}{1+j\omega R_{el}C_{el}} + R_{gel} + \frac{R_{ep}}{1+j\omega R_{ep}C_{ep}} + R_d \tag{3}$$

E_{hc} is the half-cell electrode/gel potential, C_{el} is modeled accumulated charge between the electrode and the skin, R_{el} is the resistance between the electrode and the skin during charge transfer, and R_{gel} is the resistance of conductive electrolyte (gel).

THORAX IMPEDANCE

The impedance of the thorax consists of extracellular space and intracellular fluid. It has a frequency dependence, ie in the low frequency range (1-70 kHz) the capacity of the intracellular fluid has a large value of impedance, and electric current flows mainly through the extracellular fluid, and in the high frequency range (70 kHz - 1 MHz), when the impedance of the intracellular fluid decreases, the current flows through both the intracellular fluid and the extracellular fluid, depending on their relative conductivity and volume [9]. The electrical equivalent circuit of the thorax

tissue consists of a resistor representing the extracellular space and a parallel resistor representing the intracellular space connected in series with a capacitor simulating the insulating properties of the cell membrane, which is schematically shown in Figure 3.
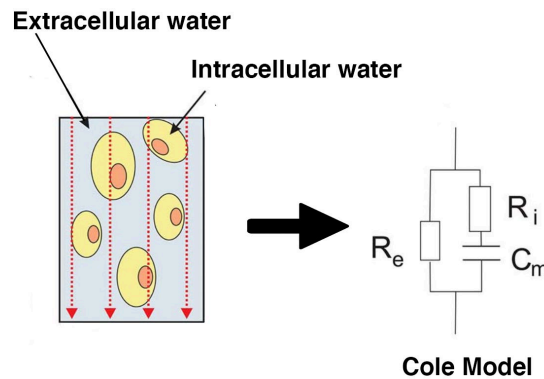

Figure 3. Electrical equivalent circuit of current flow through intracellular and extracellular fluid

Cole's model is valid for measuring body parts like thorax [10]. Since two pairs of conductive $i(j\omega)$ and two pairs of measuring electrodes $u(j\omega)$ are used in the measurement, the total resistance can be calculated according to Ohm's law (5):

$$\bar{Z} = \frac{\bar{u}(j\omega)}{\bar{i}(j\omega)}.$$

(4)

Using the Cole model, the frequency resistance is described as follows:

$$\bar{Z} = \frac{\bar{u}(j\omega)}{\bar{i}(j\omega)} = \frac{R_e \cdot (R_i + \frac{1}{j\omega C_m})}{R_e + R_i + \frac{1}{j\omega C_m}} = (\frac{R_e}{R_i + R_e}) \cdot (R_i + \frac{R_e}{1 + j\omega C_m(R_i + R_e)})$$

(5)

with the following real and imaginary part:

$$\text{Re}\{Z\} = \frac{R_e + \omega^2 + C_m^2 R_i R_e(R_i + R_e)}{1 + \omega^2 C_m^2(R_i + R_e)^2},$$

(6)

$$\text{Im}\{Z\} = -\frac{\omega C_m R_e^2}{1 + \omega^2 C_m^2(R_i + R_e)^2}.$$

(7)

It should be noted that since only the absolute value is measured at a fixed frequency, the equation for the base resistance will look like:

$$Z_0(\omega) = |\bar{Z}(j\omega)| = \sqrt{\frac{R_e^2 + R_e^2 R_i \omega^2}{1 + C_m^2 \omega^2(R_i + R_e)^2}}.$$

(8)

EXPERIMENTAL COMPLEX

To test and calibrate measuring devices for measuring hemodynamic parameters based on changes in bioimpedance, an experimental complex was developed that reproduces the measurement of bioimpedance by modeling electrical processes of the human body, specifically thorax, in the form of an electrical equivalent circuit.

A Cole model was used to represent the thorax impedance. Below is a schematic representation of places of application of electrodes (Fig. 4) for one measurement channel and the electrical equivalent circuit of the chest to measure the change in

bioimpedance for one channel (Fig. 5), for the left part of the thorax measurement scheme will be similar.
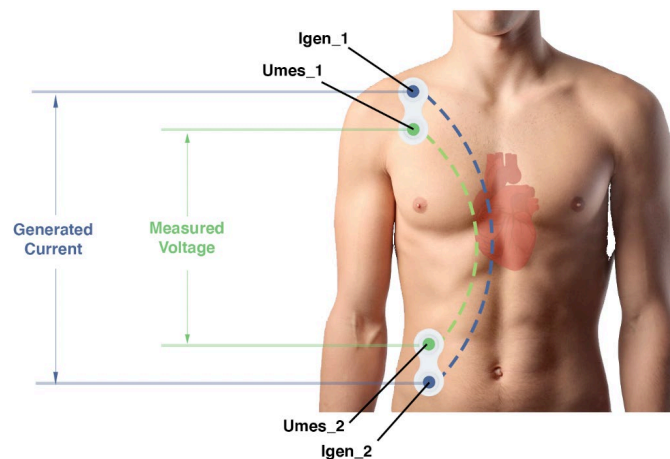

Figure 4. Schematic representation of the location of the electrodes on the chest.

In the simulation should take into account the need for possible changes in baseline impedance for signal simulation capabilities for patients with different height and body composition [11]. To simulate the base impedance and impedance changes during time heart rate using variable resistance R_e, for simulate of patient-contact electrode using variable resistance Z_{el}. Since an ECG signal is required to determine the reference points for calculating hemodynamic parameters by measuring the changes of biompedance, an ECG signal generator was used - U_{ecg}.
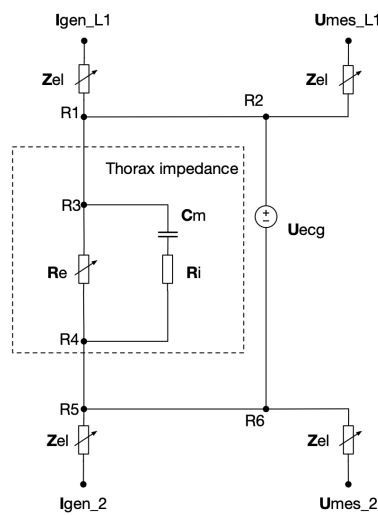

Figure 5. Electrical equivalent circuit of the chest to measure changes in bioimpedance.

To implement a dynamic change for bioimpedance Re, was implemented a circuit from digital two-channel potentiometer AD8402, controlled by a microcontroller. This scheme allowed to obtain the necessary combinations of resistance in the range from 50Ω to 1000Ω, with adjustment resolution of 3.9Ω. The connection diagram of the digital potentiometer for the two channels (right and left) is shown in Figure 6.
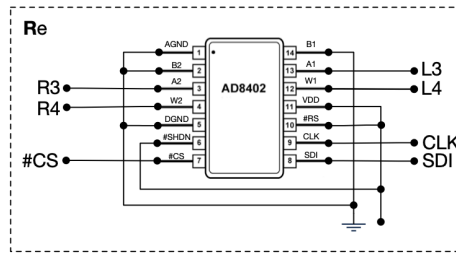
Figure 6. Wiring diagram AD8402

The appropriate range of resistance combination was calculated based on the measured values of the change in the impedance of the human thorax. Variable resistance includes the basic impedance of the thorax R_0, adjustable to human anthropometric features such as height and weight and dynamic change under the influence of cardiac output R_e, the equation of dynamic resistance of the thorax is given below (9):

$$Re = R_0 + \Delta R_e .\qquad\qquad(9)$$

To implement the interface impedance of the electrode Z_{el}, an RC circuit with variable resistors was used, due to which it is possible to reproduce situations with non-ideal contact patient-electrode. Such situations in clinical practice are frequent and possible while conditions of scin is not proper: sweating, the presence of hair in the places of application of electrodes or the use of expired electrodes. The Z_{ell} scheme is shown in Figure 7.
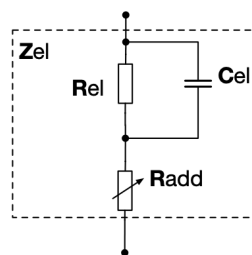

Figure 7. Shematic of Z_{el}

Also, an ECG generator is connected in parallel with the thoracal impedance circuit, as most bioimpedance-based hemodynamic devices use an ECG curve to find reference points for calculation hemodynamic parameters. The ECG signal generator is implemented on the basis of a 12-bit digital-analog output of the controller and two built-in microcontroller operational amplifiers. This scheme will generate a signal with an amplitude up to 2 mV. Since the control program is same for changing the impedance and formation of the ECG signal, this ensures the consistency of the generated signals. STM32L15RE was used as the CPU, Figure 8 shows a simplified diagram of the hardware of the experimental complex. The ECG signal uses 2 channel 12-bit DAC with output buffers, the SPI interface is used to control the digital rheostat, and the UART protocol is used to communicate with the PC with control program.

The software part of the complex consists of an exchange protocol, resistance recalculation algorithms and an ECG signal generation algorithm. A simplified diagram of the software is shown in Figure 9.
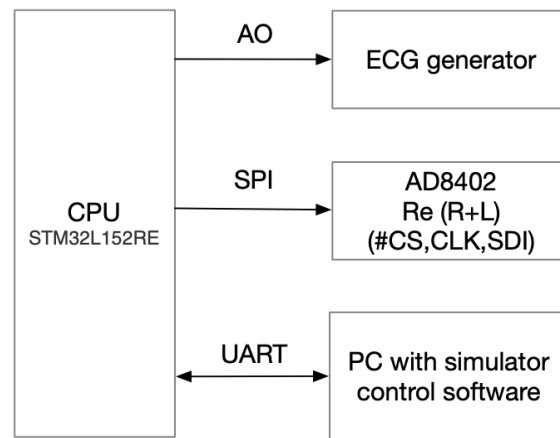
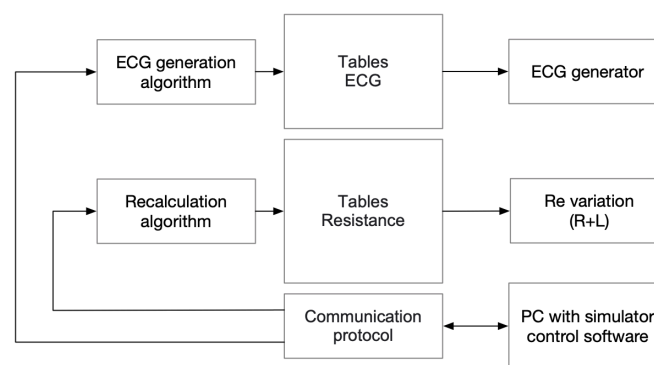**Figure 8. Simplified scheme of the experimental hardware**

Figure 9. Simplified scheme of the experimental software

RESEARCH RESULTS

The study substantiated to form the basic requirements for the creation of an experimental complex and offers hardware and software solutions for comprehensive verification and calibration of measuring instruments of central hemodynamic parameters based on changes of bioimpedance. The proposed solutions allow to simulate the operation of the measuring device based on the change of bioimedance in the area of the human body such as the thorax.

These solutions also eliminate possible limitation in the bioimpedance simulators proposed by other researchers, such as a digital potentiometer can create a signal of change of thorax impedance a certain shape smoothly and with a clear and understandable discreteness of 3.9 Ω.

CONCLUSION

The proposed complex will allow testing of existing measuring devices, as well as allow to validation and conduct experiments while developing new technical solutions and methods based on measurements of bioimpedance changes. For further development of the complex, it is necessary to take into account the possibility of imitation of artifacts - motor, respiratory, effects of electrosurgical instruments, because under such conditions significantly worsens the correlation of the bioimpedance method compared to the "gold standard" method of thermodilution. It is necessary to implement modeling of these artifacts in the further development of the experimental complex.

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Анотація. В даній статті проведено дослідження умов та розробка експериментального комплексу для перевірки та калібрування вимірювальних пристроїв параметрів центральної гемодинаміки на основі зміни біоімпедансу. В якості найбільш перспективного та широко застосованого методу обрано методі імпедансної кардіографії, що полягає у пропусканні слабкострумового (0,5...5 мА) високочастотного (20-200 кГц) сигналу через грудну клітку, та вимірювання падіння напруги, що обумовлене зміною біоімпедансу грудної клітки під час викиду крові серцем. Було досліджено можливість моделювання імпедансу грудної клітки, та на підставі отриманих результатів було розроблено електричну еквівалентну схему. Розроблено апаратну та програмну частину експериментального комплексу, що дозволяє моделювати базовий імпеданс грудної клітки та його динамічну зміну під час викиду крові серцем у діапазоні від 50Ω до 1000Ω. Також у складі комплексу передбачено генератор сигналу ЕКГ, так як більшість вимірювальних пристроїв даного методу використовують криву сигналу ЕКГ для пошуку опорних точок для подальших розрахунків.

Ключові слова: гемодинаміка, серцевий викид, моніторинг параметрів, торакальний електричний біоімпеданс.

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