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Computer-Based Vectorcardiograph for Research Purposes

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***Abstract*—The vectorcardiography (VCG) represents the cardiac electric field by vectors and provides a better morphological interpretation of the cardiac electrical field. The main objective of this paper has been to describe our VCG device that has been determined for scientific research and for the clinical diagnosis of heart diseases. We have designed the hardware as a small portable device and the software that measures data saved in a raw form accessible for further processing. The main contribution is in the new way of common mode reduction (driven right leg circuit). The designed device has a great potential thanks to the new methods of interpretation and automatic signal evaluation an also thanks to the small procurement price.**

***Index Terms*—Vectorcardiography, Frank's leads, driven right leg circuit.**

I. INTRODUCTION

Recently, a great progress in the treatment of cardiovascular illnesses has been observed. A fast and accurate diagnosis of these illnesses could reduce the number of sudden deaths. The Electrocardiography (ECG) is the most common diagnostic method, but the results are usually difficult to interpret. The ECG includes the vectorcardiography (VCG) that provides a better morphological interpretation of the cardiac electric field.

In 1913, Einthoven, Fahr and de Waart wanted to represent the cardiac electric field by vectors that have sense, magnitude and direction. Since then, the development of VCG has gone in different directions. The VCG presents several types of lead systems, for example, Frank, McFee-Parungao, and SVEC III. These lead systems differ by the number of electrodes and by spatial sensitivity properties. The most common VCG method is the Frank's reference vectorcardiography system created by Ernst Frank in 1954.

This system uses seven special located electrodes. Location of the electrodes ensures measurement independence from the heart position and body size of the examined. Five electrodes are placed on the same transverse plane (Fig. 1). The placement of the electrodes allows to record cardiac heart activity in the three orthogonal planes x, y, and z [1], [2].

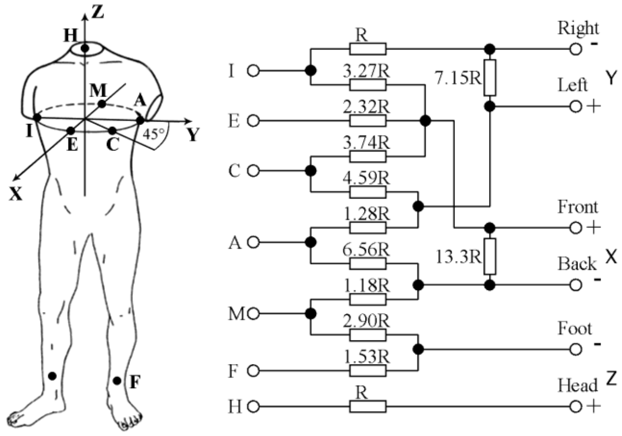

Fig. 1. The lead matrix of the Frank VCG system. The transverse plane contains the electrode E – on the front side, M - in the middle of the dorsal side, I - on the right side under the armpit, A - on the left side under the armpit. The electrode C makes an angle of 45 degrees between the electrodes E and A. The electrode H is placed on the back of the neck. The electrode F can be found on the left leg, between the knee and ankle [1], [3].

A resistor network compensates the unbalanced heart location and provides the same impedance value in every amplifier lead. The great advantage of this system is orthonormality that provides equal voltage levels of the measured X, Y, and Z leads [1].

II. PROBLEM DEFINITION

The commercially available VCGs are designed as closed systems. The specialized software of these VCGs can provide only predefined diagnostics. The data are pre processed without the possibility of the signal post processing. VCGs are often combined with conventional 12-channel ECG. The orthogonal leads can be computed from six precordial and two limb leads, or these systems display only the monocardiograms. The price of these devices can exceed the costs of common ECGs. The advantages of these devices are the complexity, functionality and the automatic diagnostics. These medical diagnostic devices are assigned to the ambulant examination of patients.

Our proposed device differs from the commercially

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available VCGs; our system provides the output data in a raw form that can be post processed. Data from the VCG are filtered within the range between 0.5 Hz and 150 Hz. We have developed the software in the LabVIEW software environment that can insert data into a database. The VCG has a better diagnostic value than the conventional ECG. Our system has been designed to expand the application of the VCG in the diagnosis of heart diseases, for example, the localization of myocardial infarction, the left anterior fascicular block and the left septal fascicular block. The VCG can also be used in the evaluation of electrically inactive areas [4].

III. THE SUGGESTED MEASUREMENT VECTORCARDIOGRAPHY UNIT

The concept of the Frank's corrected orthogonal lead system is well known [3]. The designed device consists of the Frank's lead system, a driven right leg circuit, a resistor network, instrumentation amplifiers, filters, and a data acquisition unit (Fig. 2).

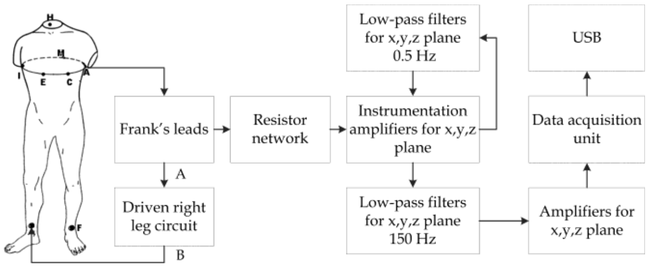

Fig. 2. Block diagram of the functional vectorcardiograph unit.

In general, the amplitude of the signals measured from the body surface takes values within the range of hundreds of µV to tens of mV. These signals are usually affected by man-made interference. There are four ways in which the electromagnetic field interferes with the measured biological signals. The interference paths are magnetic induction, the displacement current induced into the leads, the displacement current induced into the patient causing interference voltage between the two recording electrodes, and the displacement current induced into the patient causing interference voltage between the recording electrodes and the amplifier common, i.e. common mode voltage (Fig. 3) [5].

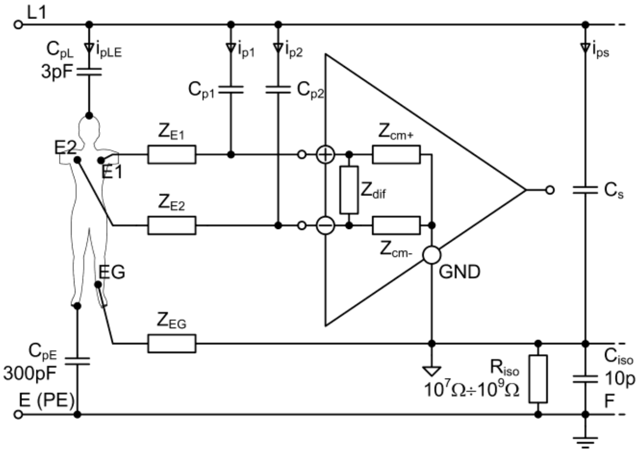

Fig. 3. Interference paths in biosignal measurements from the body surface [6].

The solutions of these four interferences are as follows. The use of mutually twisted wires reduces the interference generated by the magnetic induction.

The induced current in the *E1* and *E2* leads can be minimized by shielded wires that reduce the capacity of capacitors C_{p1} and C_{p2}. Careful electrode positioning avoids recording of the voltage caused by displacement currents flowing through the body impedance [7]. The common mode voltage of the electrodes can be removed using grounding or the driven right leg circuit. In ideal conditions, the common mode voltage is suppressed for the amplifier with a very large common mode rejection ratio (CMRR). In practice, voltage is indirectly manifested due to the unequal impedance of the leads as interfered differential voltage. It can be calculated by the following expression [5]

$$v_i = v_c\left(\frac{1}{CMMR} + \frac{Z_d}{Z_c}\right), \tag{1}$$

where v_i is the interfered differential voltage, v_c is the common mode voltage, Z_d is the difference of the impedance between two electrodes, and Z_c is the common mode impedance. *CMRR* is a parameter of the differential amplifier that represents the ratio of the power of differential gain over the common mode gain. It is usually measured in positive decibels, for real circuits within the range 60 to 120 dB [5].

As it is not possible to provide the same impedance on all leads, the value of the common mode voltage v_c has to be reduced. In our device, we have used the driven right leg circuit (an electrode on the patient's right leg) that is usually used for the conventional ECG. It provides a connection between the patient and a common amplifier in order to minimize the common mode voltage [8], [9]. The signals of the individual leads are averaged, inverted and amplified. The conduction of a signal modulated in this way to the patient provides noise reduction (Fig. 4).

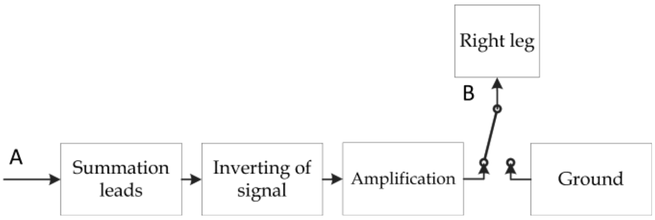

Fig. 4. Block diagram of the functional driven right leg circuit.

In the measurement of biological signals, there are more kinds of the common mode noise such as low frequency differential signals. They are located in the frequency range below 0.5 Hz and mainly produced by moving artefacts, electrodes polarizing voltage, impedance changes due to breathing, etc. The High Pass Filter (HPF) can reduce these artefacts. In our device, we use the Low Pass Filter (LPF) connected, in the feedback loop, to the instrumentation amplifier. Low frequencies can be thus simply removed.

Other interferences are electrical activity produced by skeletal muscles, noise components, higher harmonic frequencies, etc. The upper cut-off frequency equal to 150 Hz is considered as optimal for the purpose of ECG diagnostics.

The last part of our device is the data acquisition unit (DAU). The DAU consists of a 16 bit microcontroller (MCU). The MCU reads data from an analogue to digital converter (ADC) via a serial peripheral interface. We used the analogue to digital converter with a sampling frequency of 2 kHz and 12 bit resolution. The acquired data were formatted into a data packet and sent to a workstation by USB for the subsequent processing and evaluation.

The device is powered by a USB source voltage of 5 V. The patient is electrically connected to the voltage source (driven by the right leg circuit) and it is thus necessary to ensure the safety of the equipment. It means that the device has been designed in such a way that the requirements on medical devices are satisfied. One of these conditions is the galvanic isolation of the circuits connected to the patient from other electronic circuits. In our device, a galvanic isolation of the power supply and a signal bus between the ADC and the MCU buses has been made.

IV. CIRCUIT SIMULATION

Before the construction of our VCG device, we simulated the whole circuit design. The simulation was provided in OrCAD PSpice 10.5. The simulated frequency response characteristics are shown in Fig. 5. The fall of 3 dB corresponds to the frequency range from 0.5 Hz to 150.09 Hz. The band pass phase is around 180 degrees.

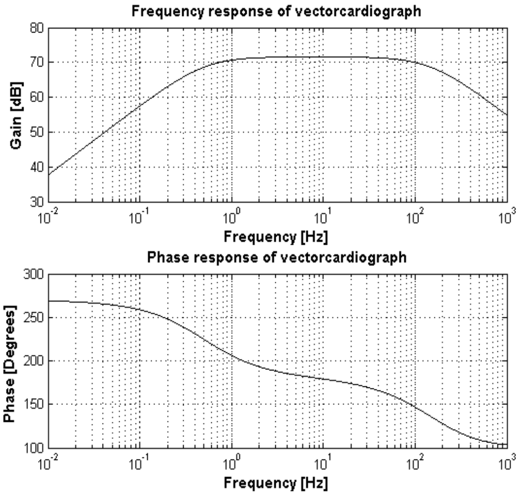

Fig. 5. Frequency and phase response of the VCG simulation.

The second simulation was to show the importance of the use of the driven right leg circuit. The simulation parameters were as follows: the source of the common mode noise with the frequency of 50 Hz was connected through the capacitive coupling 3 pF on each electrode, two sources of square waves with the amplitude 1 mV and frequency of 2 Hz were connected through the resistance of 100 kΩ, and the individual leads were separated by the 100 kΩ resistance. The result of the virtual grounding is shown in Fig. 6(A).

The virtual grounding corresponds to the grounding of the patient. The amplitude noise in our simulation was 25.4 mV and this value was not negligible due to the signal characteristic. In the last simulation, the ground electrode was connected to the driven right leg circuit.

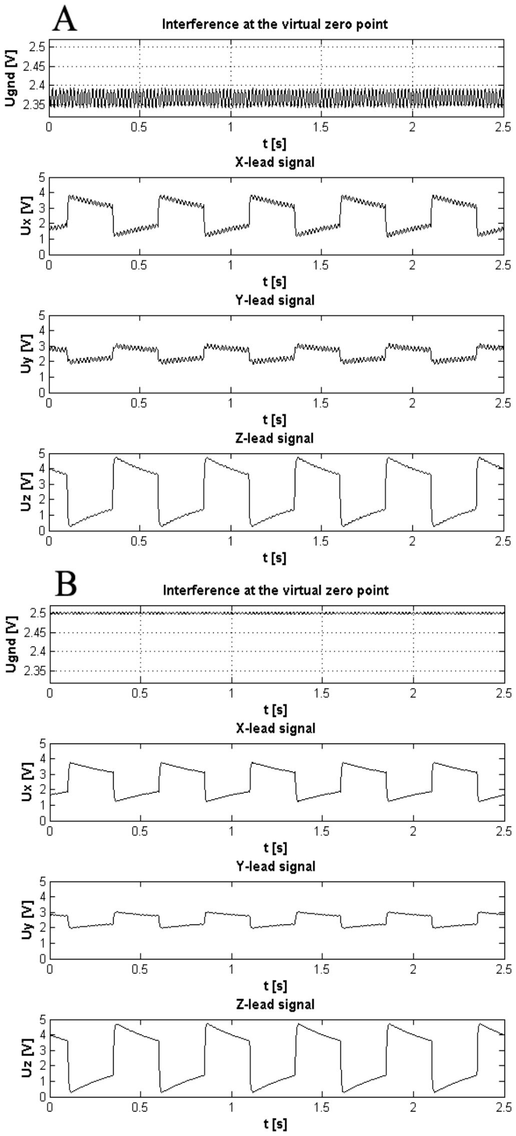

Fig. 6. Simulation of the VCG in PSpice without the driven right leg circuit (A) and with the driven right leg circuit (B).

The results showed a significant improvement (Fig. 6(B)). The amplitude noise was reduced from 25.4 mV to 3.4 mV.

V. DEVICE IMPLEMENTATION

The VCG was constructed as a small portable device (15 cm × 8 cm) powered by USB. The analogue and digital parts have been on separated printed circuit boards. These parts have separated grounds which are connected only in one point in the circuit.

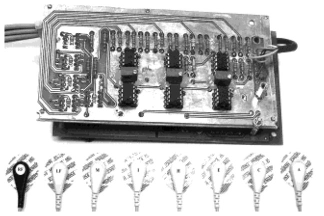

Fig. 7. VCG hardware realization with outlet external leads.

The device has been inserted into a plastic box with external outlet leads, a USB connector, and LED. The LED indicates that the device is in recording mode. The uncovered device is shown in Fig. 7.

VI. MEASUREMENTS AND TESTS

Our proposed device was tested in natural conditions with the same input signal as during the simulation part. For the generation of a square wave, the ECG generator Fluke 7000DP was used. No other additional software filtering was performed on the output data. The results of the measurement are shown in Fig. 8. The shape of the signal was affected by low-frequency noise suppression.

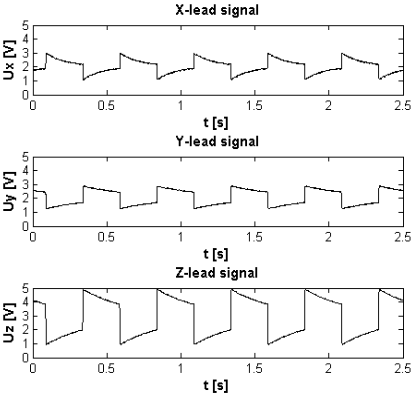

Fig. 8. The results of the test with the ECG generator Fluke 7000DP in natural conditions.

VII. GRAPHIC USER INTERFACE AND DATA INTERPRETATION

In order to demonstrate the output data, we have developed a simple graphic user interface (GUI) based on the LabVIEW software (Fig. 9). The GUI allows you to view the VCG curve as the three one-dimensional signals in time.

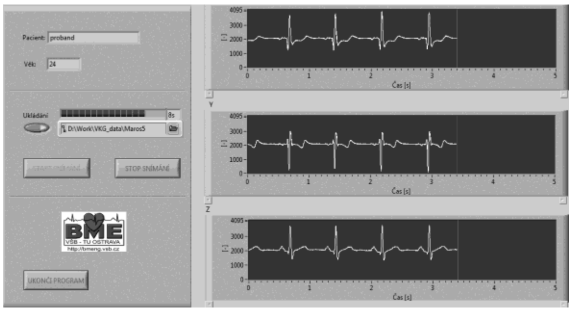

Fig. 9. The GUI with measured native VCG signals in the LabVIEW environment.

The main objective of the proposed software is to capture the data of electrical heart activity and save them in a database. The data are saved in the database with identification of the examined that allows a retroactive interpretation of the results. The GUI controls operations such as starting, stopping the recording and also saving the records.

The output signals represent the orthogonal planes x, y, and z. These signals can be displayed in different views (1D, 2D and 3D).

The one-dimensional view of the signals does not provide new information about the heart activity. The 3D view performs the heart vector movement. The diagnostic importance has the representation of dependencies between the individual planes (the 2D and 3D view).

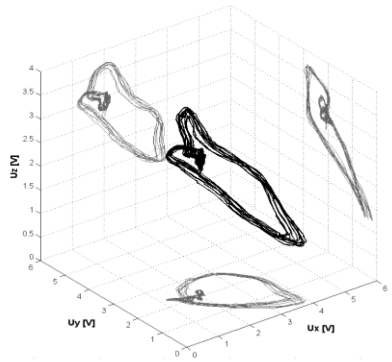

Fig. 10. 3D view with the individual 2D projections (10 heart cycles).

The heart electrical activity (atrial depolarization, ventricular depolarization, atrial repolarization, and ventricular repolarization) is represented there by loops (Fig. 10). Their shapes' changes can provide information about potential pathology.

VIII. CONCLUSIONS

The objective of this work has been to propose, implement and test VCG for scientific research, the clinical diagnosis of heart diseases, and for education. We proposed both hardware and software parts of this equipment. We see the main contribution in the new way of common mode reduction. Measured data saved in raw form are easily readable and accessible for further processing. The great potential of our designed device is seen in the new methods of interpretation and automatic signal evaluation. Our device is more affordable than the vectorcardiographs with computed data interpretation.

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