COMBINED USE OF ELECTROCARDIOGRAPHY AND ACCELEROMETRY TO EVALUATE THE PHYSICAL CONDITION OF A SUBJECT

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Abstract: The advances in the technology of integrated circuits with low voltage bias allow the development of low-cost, low-power and downsized portable instrumentation. Regarding biomedical applications, mobile medical sensors provide an efficient and accurate way for evaluating the physical condition of a subject, without the need of direct intervention of a healthcare professional. The proposed equipment combines both electrocardiography and accelerometry, allowing the correlation of mobility with cardiac events in real-time. The detailed analysis of the implemented electrocardiograph circuit is presented. The most relevant experimental results obtained with the proposed instrument are shown. The results have proven that the device is able to generate low-noise signals and to provide a precise tracing of the cardiac cycle, in comparison to other equipments used for clinical applications. Further optimization is needed, specifically focusing the implementation of the wireless transmission of the gathered data and the miniaturization of the system.

1 INTRODUCTION

Monitoring of patients' health is a topic that will benefit from the use of portable devices, reducing costs and improving healthcare systems quality. Long-term applications of portable devices include management of cardiac/respiratory diseases and assistive technology for the elderly, being also relevant in the field of rehabilitation. Moreover, they can also be useful to study the progress of athletes' performance [1].

Electrocardiography (ECG) is a simple, fast and accurate method for the early detection of cardiac pathologies, through the analysis of heart rate and interpretation of the tracing of the cardiac cycle [2]. Continuous monitoring of the heart's electrical activity during daily life provides information about the response of the cardiovascular system to different types of activities. Therefore, the combination of ECG signals and body motion information, such as the one given by accelerometers, is a way to correlate cardiac events and mobility, allowing the evaluation of the subjects' physical condition. Furthermore, accelerometers have been shown to be an objective and reliable tool to estimate energy expenditure [3, 4].

The main purpose of this work was to develop a portable instrument that combined information from electrocardiography and accelerometry. Thus, an electrocardiograph prototype circuit was set up and integrated with a 3-axis accelerometer (ADXL345). Additionally, data acquisition hardware was used (Arduino Uno) and a software interface was developed in MATLAB, which allowed analysis and storage of physiological data in a computer. The interface also enabled the calculation of speed, distance covered, energy expended by the subject and heart rate.

2 METHODOLOGY

2.1 System Setup

The electrocardiograph circuit was simulated in a general-purpose circuit simulation program (5Spice). The main circuit can be divided into 5 sub-circuits. Each one will be analysed separately. The functional diagram is presented in figure 1.

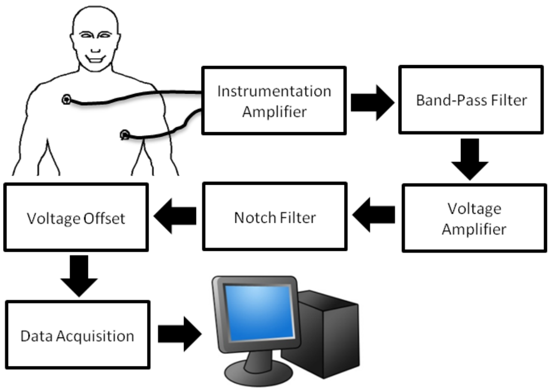

Figure 1: Functional diagram of the electrocardiograph.

The first sub-circuit, represented in figure 2, is responsible for the acquisition and amplification of the low amplitude ECG signal that lies in the millivolts range. The typical design approach is to use an instrumentation amplifier (IA) that multiplies the difference in the two inputs, reducing common-mode noise. For this type of application it is required a low-noise and low-power consumption IA. The INA129P amplifier is well suited for this purpose, being a device widely applied in medical instrumentation [5].

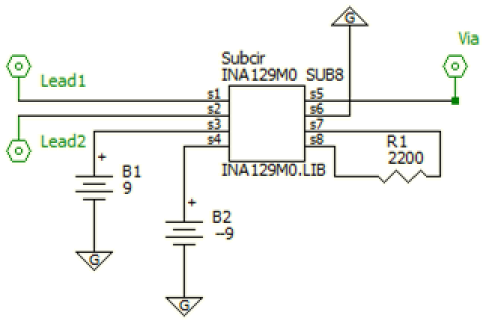

Figure 2: Instrumentation Amplifier circuit schematic.

The INA129P gain (G) is given by equation 1:

$$G = 1 + \frac{49.4\ k\Omega}{R_1} \quad (1)$$

The gain was adjusted by an external resistor, R_1, and was set to 23.5 in the present application.

The electrodes, with high output impedance, interface directly with the inputs of the IA. It is advisable the use of a conductive gel to further reduce the electrical impedance.

The sub-circuit following the amplifier is a band-pass filter (BPF) transparent to the frequency band of interest, which for electrocardiographic events is in the range of 0.05-250 Hz, although standard ECG clinical application suggests as sufficient a bandwidth of 0.05 Hz to 100 Hz. For simplicity, the BPF was designed with discrete components: a passive low-pass filter followed by a passive high-pass filter. The chosen cut-off frequencies are 0.05 Hz and 106 Hz, respectively, for the high and low-pass filter. The transfer function H(s) of the filter is given by equation 2:

$$H(s) = \frac{667s}{s^2+667s+202} \quad (2)$$

The schematic diagram and the simulated output of the band-pass filter are depicted below.

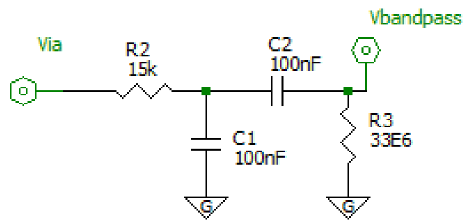

Figure 3: Band-pass filter circuit schematic.

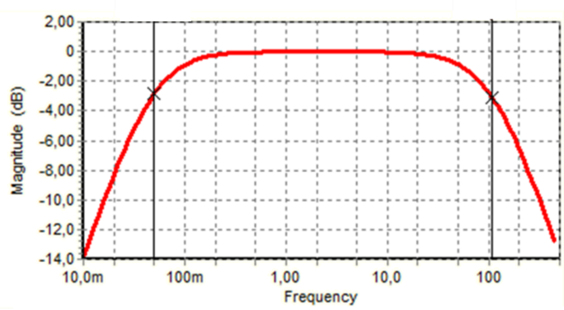

Figure 4: Simulation of band-pass filter behaviour. The cut-off frequencies are 0.05 Hz and 106 Hz.

The signal is amplified, once more, in the third sub-circuit; the variable gain can be set between 69 and 169. A low offset voltage and low-noise operational amplifier (op-amp) was chosen (LF412) [6]. The schematic of the circuit is presented in figure 5.

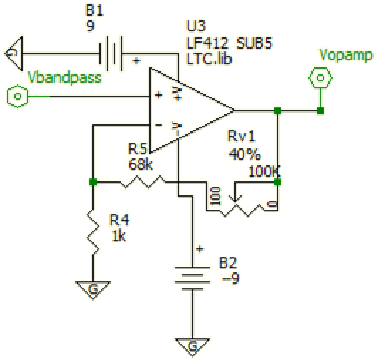

Figure 5: Schematic of the second amplifier.

The next circuit module is a notch-filter (NF), designed to attenuate the electric grid noise at 50 Hz. For this purpose, a band-stop filter centered at 48 Hz was implemented according to the schematic below. Its alternating current (AC) simulation is depicted in figure 7.

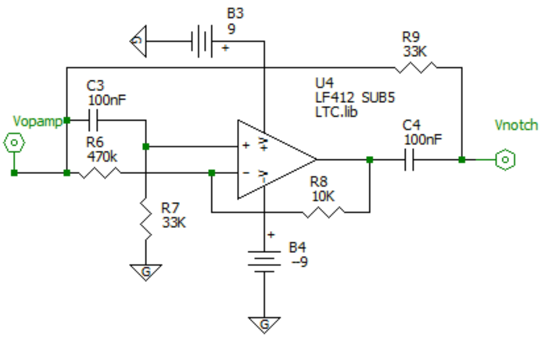

Figure 6: Active notch filter circuit schematic.

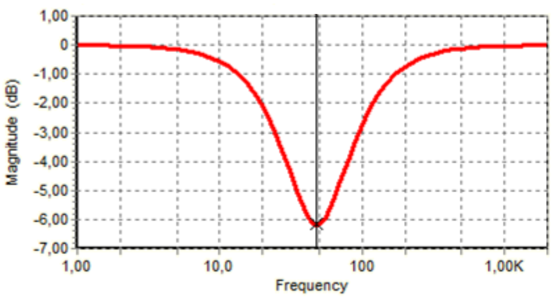

Figure 7: Simulation of notch filter behaviour. At 48 Hz, the central frequency, the loss is approximately 6 dB.

The transfer function of the notch filter is given by equation 3:

$$H(s) = 0.5 \times \frac{s^2+302^2}{s^2+439s+302^2} \qquad (3)$$

Since signals between 0 and 5 V are required for the output of the electrocardiograph, in order to be used as an input of the selected Analogue to Digital Converter (ADC), belonging to the Arduino Uno, a voltage offset circuit was implemented. The picture below (figure 8) shows its electrical schematic.

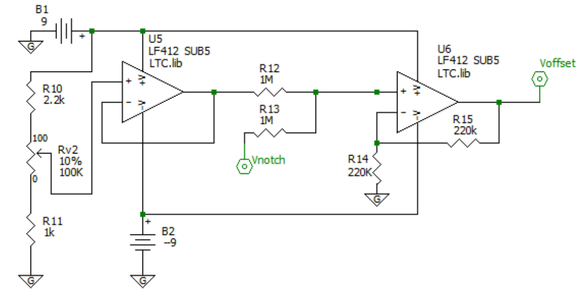

Figure 8: Voltage offset circuit.

The voltage shift can be adjusted by the voltage divider, to any value between 0 and 9 V. The first op-amp is used as a buffer to avoid impedance loading and to provide low impedance reference, whereas the second one acts as a signal adder.

Another circuit was added to provide the user with feedback about the battery status (figure 9). Thus, a Light Emission Diode (LED) will emit light when the battery level lowers approximately to 6.2 V. The circuit consists of a comparator with a voltage reference of 2.7 V, set by a Zener diode.

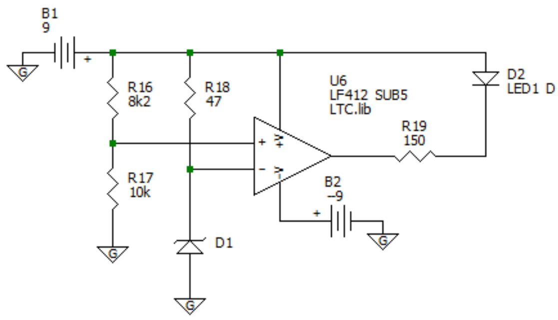

Figure 9: Circuit which monitors the battery.

After simulating and validating the aforementioned modules, the circuit was implemented in a stripboard. The layout was made using EAGLE software.

The electrocardiograph circuit was also integrated with a 3 axis-accelerometer (ADXL345). The outputs of the electrocardiograph and the accelerometer were connected to a microcontroller platform, the Arduino Uno, which acquires analogue signals and converts them into digital signals and sends them to a computer via Universal Serial Bus (USB) link. For that, the necessary codes were uploaded to the board through the open source Arduino Software. The system setup is depicted in figure 10.

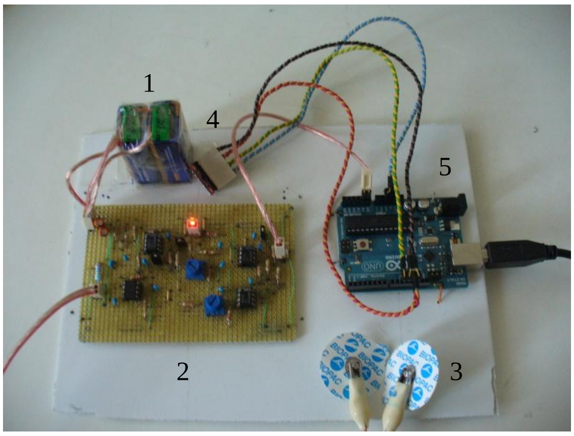

Figure 10: Setup of the complete system. 1 – 9 V battery supply; 2 – electrocardiograph; 3 – disposable electrodes; 4 – accelerometer; 5 – data acquisition hardware, Arduino Uno.

2.2 Signal Acquisition and Parameter Estimation

The ECG signals were obtained with electrodes attached to the subject's chest, positioned according to an appropriate derivation. Care was also taken to set the signals ground in a neutral (non-electrical conductive) body region, such as the prominent bony area of the wrist or the clavicle.

The accelerometry data were acquired by holding the accelerometer in a fixed position in the trunk of the subject, in such a way that the z-axis would match the direction of the movement.

User-friendly software interface was developed in MATLAB, allowing in real-time the simultaneous acquisition of both signals every 18 ms. The interface was also able to calculate speed, distance covered and heart rate. Furthermore, the signal from the accelerometer allowed the estimation of the energy expenditure. Each parameter was updated every 5 seconds.

Previously to the estimation of the parameters, the signal from the accelerometer was filtered by a windowed linear-phase FIR (Finite Impulse Response) digital filter.

The speed was estimated according to equation 4.

$$Speed = steps\ per\ 2\ s\ \times \frac{stride}{2\ s} \qquad (4)$$

A step was defined as the occurrence of a negative slope in the acceleration plot in the z-axis. In particular, the crossing of a dynamic threshold by the acceleration curve was used to identify if an effective step had been taken. The peak detection algorithm considered a local minimum as a step if the time interval between the actual and previous step was higher than two times the standard deviation of the accelerometer signal. Experimental data in [7] were used to evaluate the stride, which depends on the height of the subject and the number of steps computed per two seconds. The distance covered is simply calculated by multiplying the speed by the window length (5 seconds).

The energy expenditure (EE) is obtained from the linear regression in equation 5, derived from experimental data in [8], in which body mass was used to predict the calories burnt per minute at walking speeds between 3.2 and 6.4 km/h.

$$EE\ (Kcal/min) = -2.494 + 0.047 \times Weight\ (kg) + 0.729 \times Speed\ (Km/h) \qquad (5)$$

The peak detection algorithm previously described is also useful for the identification of the R-peak in the ECG data, allowing the estimation of the heart rate.

3 RESULTS

In figure 11, the most relevant results are shown.

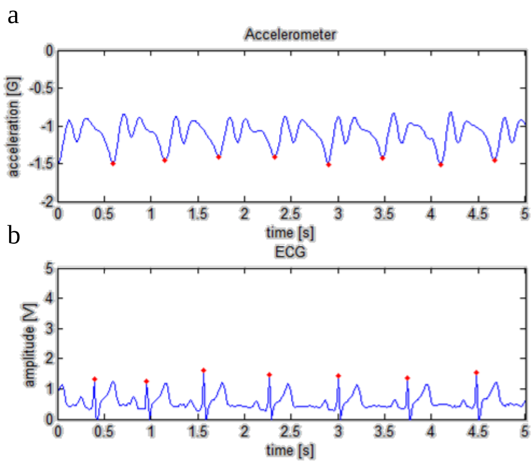

Figure 11: a – Data from the accelerometer (z-axis), where the red dots mark the steps during walking; b – Data from the electrocardiograph, where red dots mark the heart beats.

The body motion (figure 11a) and ECG signals (figure 11b) were acquired from a 22-year-old, healthy female subject (height: 1.55 m, weight: 60 Kg), walking at moderate intensity.

The electrodes were placed in a derivation equivalent to the standard lead II seen in a 12 lead ECG [9]. The calculated speed was 3 Km/h, the distance covered 125 m, the total energy expenditure 6 Kcal and the heart rate 84 beats per minute (bpm).

4 DISCUSSION

According to the results, the instrument prototype developed yields low noise ECG signals that can be efficiently analyzed with the additional software and hardware. These results have proven that it is possible to obtain a clear tracing of the cardiac cycle, with only one derivation, comparable to those obtained with a 12-lead ECG used for clinical applications. When acquired simultaneously with the accelerometry data, it enables the correlation of cardiac events with the physical activity intensity.

Further optimization is needed in the instrument, such as the implementation of the electrocardiograph circuit on a Printed Circuit Board (PCB), using a more effective ground plane, in order to reduce signal noise. For miniaturizing the system, the refactor of the layout and the use of surface mounting components are desirable. On the other hand, the addition of a memory storage device or a wireless communication system and the replacement of the Arduino Uno host-board by a dedicated low-power microprocessor would result in increased autonomy and portability. Nevertheless, the main goal of the project was accomplished, in terms of the construction of a low-cost, low-power and portable electrocardiograph/accelerometer integrated instrument for biomedical applications.

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