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Wireless ECG and Heart Rate Monitoring Using Dual Ground Dry Electrodes

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Abstract- In this work, a novel wireless easy-to-use measuring system is presented to acquire the Lead I ECG signal and the heart rate. The system presents a novel application of the dual ground configuration to dry electrodes, to reduce their level of power line interference. With this, a good quality Lead I EGC signal can be obtained simply by placing the right and left hands on the dual electrodes. Heart rate is obtained with a novel algorithm based on the continuous wavelet transform (CWT), especially designed to avoid the electromyographic noise that can be present when acquiring signals in the hands. The algorithm presented has been tested in twelve subjects of different age and physical condition, obtaining a 99.7% of sensitivity and a 100 % of positive predictivity.

I. Introduction

Recent studies [1] conclude that between the years 2010 and 2050 the population in the European Union older than 65 is expected to grow a 77%, and the population over 85 years is going to be multiplied by three. Also medical expenses will grow from 9% to 19% in 2020, and in this year there will be around 60 million people with more than 60 years. On the other hand, cardiovascular diseases are the main cause of death in ages between 44 and 64 years. All those factors have fostered the development of many novel techniques for health supervision in non-clinical environments. The use of these techniques has several benefits like allowing a more frequent supervision of patients with health troubles or also allowing patients to make part of the hospitalization at home, reducing the hospital occupancy and improving their quality of life.

Electrocardiogram (ECG) is an especially useful and widely accepted tool to detect and diagnose cardiovascular diseases. Furthermore, early detection of their symptoms is a key factor to avoid irreparable damages or even death. Nevertheless, traditional ECG acquisition systems are embarrassing and difficult to use, because they require the use of several cables and electrodes attached to the body, sometimes with conducting gel to increase the contact, and they are not prepared to transmit or store the digitalized data. Some recent works in the field have reduced some of these problems by implementing wireless ECG systems. Nevertheless, most of them [2-4] still use wet electrodes and conducting gel, whereas only few [5, 6] use other type of electrodes, mainly capacitive. Although these systems avoid some of the discomfort problems of the formers, they are designed to be worn on the thorax, hence requiring also some preparation time and skill to acquire the ECG signal. This fact represents a minor drawback for long-term monitoring but makes them unpractical for short-term or periodic monitoring for which a fast easy-to-use ECG acquisition method without any previous preparation would be highly preferable, especially for non-trained users in non-clinical environments.

In this work, we present a novel wireless system for short-term ECG acquisition in a non-clinical environment intended to be easy-to-use for non-technical users. Alternatively, their use as a fast and simple method for a first EGC acquisition in clinical environments can be also considered. The system acquires the Lead I ECG signal in monitor mode by using two dual ground dry electrodes, placed on a rigid support (for this work a plastic steering wheel has been used) from which the signal is acquired simply by placing the left and right hands on them. Dry electrodes can have a higher level of power line 50/60 Hz interference than other types of electrodes, especially in short-term measurements [7] and we propose to apply the dual ground configuration [8] to them, due to its ability to reduce 50/60 Hz interferences [9]. The system is composed by a compact battery-supplied wireless node that acquires and transmits the signal, and by a central node, connected to a personal computer to process and display the collected data or to transmit them through Internet to a medical center. The system includes also a novel algorithm to detect the heart rate based on the continuous wavelet transform (CWT), suited to the particularities of the developed system, that has been tested in twelve subjects of different age and physical condition for this work.

II. Design of the system

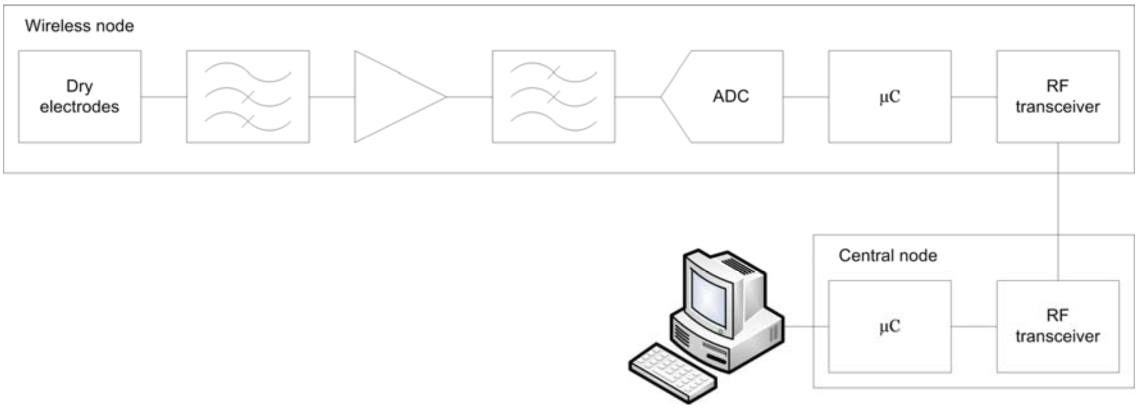

Figure 1. Block diagram of the system

Figure 1 shows a block diagram of the presented system. As it can be seen from the diagram, the wireless node acquires the Lead I ECG with the dual ground dry electrodes. An analog front end is used to amplify and filter the signal, and a microcontroller with an RF module is used also to sample and transmit the data. The central node receives the data from the wireless node and sends them to the computer using an USB port. Moreover, the central node is prepared to incorporate more than one wireless node and to handle the communication between them. Finally, the data are displayed using LabVIEW® in the same PC or transmitted through Internet. The user interface includes the implementation of the novel algorithm to detect the heart rate. In the following sections, further details of the different parts of the systems are given.

A. Wireless Node

Figure 2 shows the prototype of the wireless node developed. The dry stainless steel button electrodes are mounted on a plastic wheel in dual ground configuration [8]. This configuration achieves a greater reduction of the 50/60 Hz interference by placing a ground electrode very close to each one of the recording electrodes, instead of using the typical three electrodes configuration for the Lead I ECG, in which one ground electrode is required in the right leg. The main advantage of the configuration used is that the ECG signal is acquired simply by placing the left and right hands on the electrodes, thus avoiding any preparation procedure.

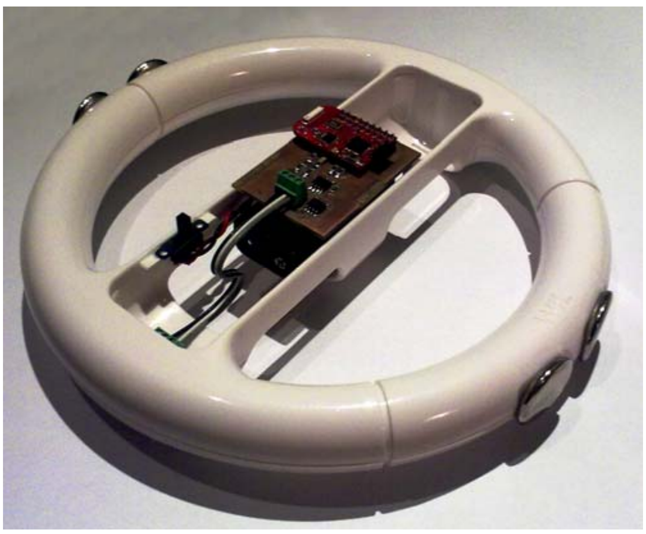

Figure 2. Wireless node prototype

All the components of the node have been designed to work with a single supply voltage of V_{cc} = 3 V supplied with batteries and the total measured current consumption of the node is 2.5 mA. With these values, and using two standard 1250 mAh 1.5 V batteries, the system will be able to perform 3000 short ECG acquisitions with a duration of 2 minutes each, which is enough for the intended short-term term monitoring purposes of the device.

The analog front-end included in the wireless node is used to filter and adapt the ECG signal level to that of the ADC of the microcontroller. The circuit is intended to acquire the Lead I ECG signal in monitor mode (frequency bandwidth between 0.5 Hz and 40 Hz [10]) and consists of several stages. First in the signal path, two buffers have been implemented, using the internal Op Amps available in the microcontroller, to reduce interferences due to the impedance mismatch of the electrodes. Then, the high-pass 0.5 Hz limit of the desired monitoring bandwidth has been achieved with a first order differential filter. This filter has been proven to yield a higher value of CMRR (Common-Mode-Rejection-Ratio) than other differential filters [11]. The differential amplifying stage has been implemented with the instrumentation amplifier INA122, which is designed to provide excellent performance in portable devices. The block gain has been set to 520. Finally, a second order low-pass filtering stage has been implemented using a Sallen-Key cell. The -3 dB corner frequency has been set to 40 Hz, which fulfills the requirements for ECG recording in monitor mode. The operational amplifier selected to implement the filter has been the low-power OPA336, which is designed for battery-powered applications. The measured CMRR of the total circuit is about 80 dB in the monitoring frequency range, mainly due to the relatively low values of CMRR of the instrumentation amplifier compared to other models. The INA122 has been designed to optimize power consumption, so other parameters are not as good as it is common in general purpose instrumentation amplifiers.

The microcontroller and the RF module of the prototype node have been implemented using the EZ430-RF2500 board that includes the MSP430F2274 microcontroller and the CC2500 transceiver. The EGC signal coming from the analog front-end was sampled with the internal 10 bits ADC of the microcontroller at a 100 Hz sampling rate that allows the easy implementation of a digital square filter of 2 taps, centered on 50 Hz to increase the rejection to power line interference.

B. Network and User Interface

The network protocol used has been SimpliciTI®, which is a Texas Instruments proprietary implementation of the IEEE 802.15.4 standard. The wireless node has been programmed to send 10 samples on each packet to minimize the power consumption due to data transmission, so the packets are sent every 100 ms in 15 bytes packets (bit rate 1200 bps), including also information about node identification and the battery level. This rate is high enough to be observed as continuous for the human perception. The central node transmits the data to a PC through the USB port configured as a serial port at 9600 bps. This allows a theoretical maximum number of up to eight active wireless nodes, which is enough to use the developed implementation in a small Wireless Body Area Network [12]. The user interface of the prototype developed has been implemented using LabVIEW® and it is used to show the acquired Lead I ECG signal and the heart rate obtained with the heart rate detection algorithm.

C. Heart Rate Detection Algorithm

The heart rate detection algorithm is based on the ECG wavelet analysis. Wavelet analysis has been applied to ECG signal with several purposes [13], being one of them obtaining the heat rate. Classical detection algorithms [14], based on adaptative QRS detection from slope, amplitude and width information, present some drawbacks like the differences on QRS frequency bands between users and the overlap of noise on the same frequency bands, whereas the more recently developed wavelet based algorithms [15, 16] overcome some of them. A new algorithm based on wavelet analysis is proposed, especially suited to the particularities of acquired signal, which are an electromyographic interference and baseline wander levels higher than traditional systems, produced by changes of the hands pressure and the movements of the patient, especially if the user press the electrodes with excessive strength. The proposed algorithm combines wavelet transforms at two different scales, one of them being sensitive to the QRS complex of the ECG and the electromyographic noise and the other being sensitive to the T wave of the ECG. This second scale is used to confirm the detection of beat when the QRS complex has a high level of noise. The Mexican Hat mother wavelet has been chosen to implement the final version of the presented algorithm, because it has shown to be the most useful in many tests performed with different ECG recordings and different mother wavelets.

The final implementation of the algorithm is configured to calculate the heart rate value every time a new data packet is received so, according to the system data rate and packet payload, it is calculated every 10 ms. First of all, a Mexican Hat based CWT at scale 1 is applied to a 10 s signal buffer. At this scale, only the highest frequency components of the signal are present, like QRS complexes and electromyographic noise, so it is especially suited to eliminate the low frequency baseline wandering. Then, a simple peak detection algorithm is applied to detect all the peaks of the obtained signal, and all those with amplitude higher than 70% of the maximum peak amplitude are automatically classified as QRS complexes. This 30% margin has been set considering that the QRS amplitude is modulated by respiration and that the electromyographic noise in the system is typically under this level.

In some cases, when the acquired signal is strongly modulated by respiration, the electromyographic peaks can be of the same amplitude than the lowest amplitude QRS complexes. To overcome this problem, the algorithm uses T waves to distinguish them. To achieve this, a Mexican Hat based CWT at scale 25 is applied to the signal buffer and all the peaks of the resulting signal (which correspond to the T peaks of the ECG) are detected. Those peaks at the scale 1 CWT with amplitude between 70% and 40% are classified as QRS if they are followed by a peak at the scale 25 CWT in the interval between 150 ms and 350 ms after the peak, which indicates a T wave. This scale is more sensitive to the baseline wandering effects than scale 1, which means that some of the T waves could be missed. Finally, all those QRS closer than 200 ms are filtered, by discarding the ones with lower amplitude.

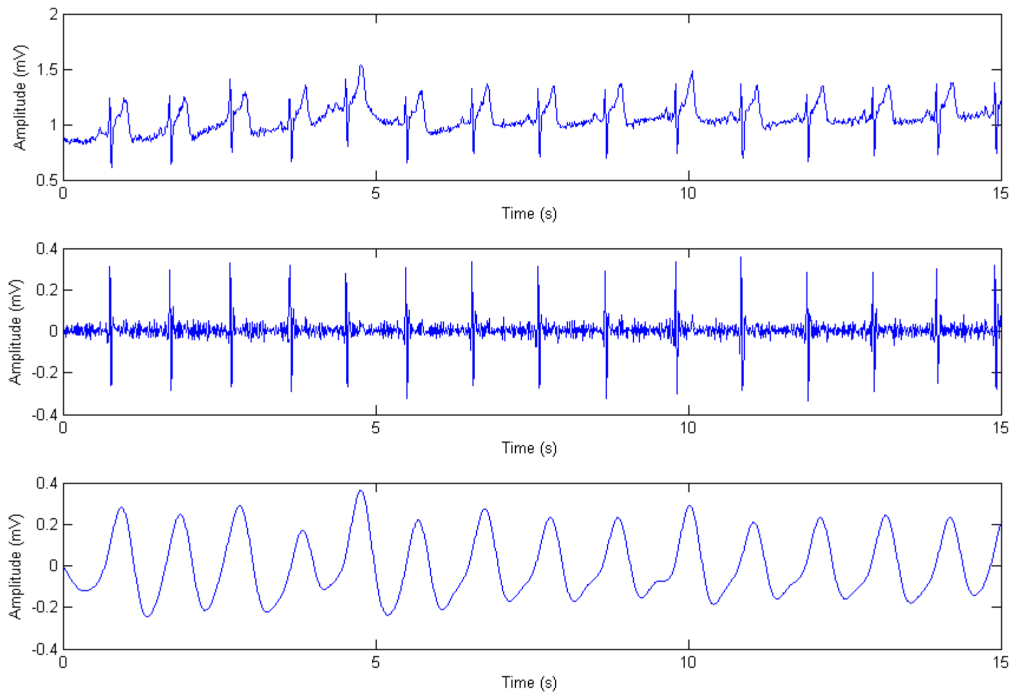

Figure 3. Lead I ECG (top), its Mexican Hat based CWT at scales 1 (center) and 25 (bottom)

Figure 3 shows a Lead I ECG signal and its associated Mexican Hat based CWT at scales 1 and 25. It can be observed how all the low frequency components of the ECG are filtered on the scale 1 signal and only the QRS complexes and electromyographic noise remain. Oppositely, on the scale 25 signal, the high frequency components of the ECG are filtered and the resulting wave has a cosine-like behavior with a peak on the same positions in which the original signal has a T wave. It can be observed that the scale 25 of the signal is slightly affected by baseline wandering.

III. Experimental Setup

In this section, the experimental protocol to characterize the ECG heart rate detection algorithm is presented. The robustness of a heart rate detection algorithm is generally characterized by two parameters: sensitivity and positive predictivity. Sensitivity is the amount of true detected beats over the real number of beats, and positive predictivity is the amount of true detected beats over the number of detected beats. Typically those parameters are calculated by using a widely accepted ECG database called MIT-BIH arrhythmia database, obtaining values over 99% on both indicators [36]. For this work, as the algorithm has been developed to specifically avoid the main characteristic sources of noise in the presented system, the test has been performed also with ECG signals acquired with the system. Test subjects have been selected to be of different age, sex, weight or physical condition. The physical condition has been divided in four groups, according to the following rules: A first group (G1) for people who make sport more than 5 days per week and follows a specific training plan or makes sport at professional level, a second group (G2) with people who usually make sport more than one day per week but without following a specific training plan, a third one (G3) with people who usually make sport one day per week and finally a fourth group (G4) with people who usually do not make sport. The test routine was to relax, sit, wait for 5 seconds to allow the system to stabilize and then perform a 60 seconds recording. The test subjects must hold the system without make excessive effort and avoid talking or moving. Table 1 shows the characteristics of the test subjects for the performance test of the heart rate detection algorithm and the results obtained.

IV. Experimental Results and Discussion

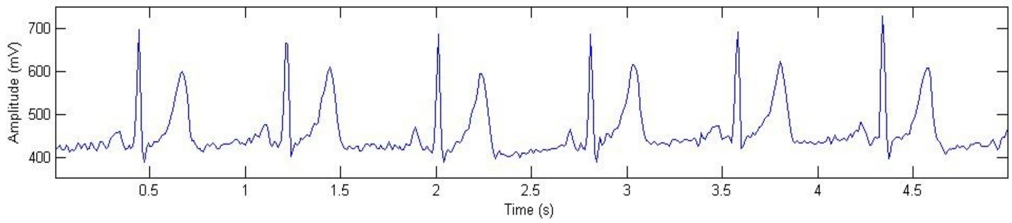

Figure 4. Lead I EGC acquired with the prototype.

Figure 4 shows an example of a typical Lead I ECG acquired with the system. It can be observed that the quality of the signal is good enough to clearly distinguish the main features and characteristic peaks of the ECG signal. From the results in Table 1, it can be observed that the heart rate detection algorithm has a good performance in terms of positive predictivity and sensitivity for the several groups of test subjects studied.

Table 1. ECG recording subjects and Heart rate detection algorithm test results

Recording	Sex	Age (years)	Weight (Kg)	Physical condition	Beats	False positive	False negative	Sensitivity	Positive predictivity
ECG 01	Male	28	73	G3	70	0	0	100%	100%
ECG 02	Male	26	74	G3	75	0	0	100%	100%
ECG 03	Female	24	80	G3	94	0	0	100%	100%
ECG 04	Male	25	67	G2	66	0	0	100%	100%
ECG 05	Female	25	53	G3	50	0	1	98.04%	100%
ECG 06	Female	56	67	G4	60	0	0	100%	100%
ECG 07	Male	23	78	G2	57	0	0	100%	100%
ECG 08	Female	20	55	G4	80	0	1	98.76%	100%
ECG 09	Female	46	58	G3	63	0	0	100%	100%
ECG 10	Female	20	65	G4	65	0	0	100%	100%
ECG 11	Male	45	93	G3	72	0	0	100%	100%
ECG 12	Male	24	60	G1	71	0	0	100%	100%
							Average	99.71%	100%

V. Conclusions

In this work, a novel wireless easy-to-use system for ECG acquisition has been presented which applies the dual ground configuration to dry electrodes to reduce their level of interference for fast short-term measurements. The system is able to capture, transmit, record and display in real time a Lead-I ECG signal, in monitor mode, and implements a novel heart rate detection algorithm suited to the particularities of the system. The algorithm presented has been validated with a group of 12 test subjects of different age and physical condition.

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