Wearable ECG System for Health and Sports Monitoring

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***Abstract*— A wireless ECG system using dry-contact electrodes is presented. The system consists of a pair of coin sized dry-contact electrodes manufactured on a standard printed circuit board that can operate both on top of the skin and through clothing and can be embedded within comfortable layers of fabric. The wireless main unit assures the transmission of the ECG signal to a computer for storage and processing. The system provides a differential gain of 56dB over a 1-150Hz bandwidth. Signals are digitized with a 10-bit ADC and samples are transferred via ultra-low power ANT+ wireless technology. The preliminary results showed that the proposed wearable system is ideally suited for monitoring patient cardiac activity in real time at home, healthcare or sports facilities. It is highly suitable for integration in wearable ECG chest harness.**

Keyword: ECG, Dry-Contact Electrode, Body Sensor

I. Introduction

Sudden and unexpected death caused by cardiac issues is an emerging health burden in the Western world. Sudden cardiac arrest (SCA) is frequently named “the silent killer”, because sudden death is often the first manifestation of cardiovascular disease. It is estimated that 2,000 people under the age of 25 die from sudden cardiac arrest in the United States every year [1]. Because of the high level of stress on the body, especially the heart, during intense competition, it is alarmingly frequent for athletes to suffer from heat stroke or fall victim of heart failure. Thus, identification of apparently normal persons who actually are at elevated risk for sudden death is a major challenge. It was found that sudden death is more likely to occur if abnormal heart rate profiles are present during exercise and recovery. However, identifying those changes require monitoring of the parameters in real time during extensive activities. We address those needs by proposing a wearable ECG monitoring system that can sense ECG signal both on top of the skin and through clothes or hair and transmit it to a computer for monitoring or logging. Because the wearable system constantly monitors the athlete during the activity, the athlete, coaches and medical professionals will be alerted if overheating or SCA indicators are detected, allowing timely intervention thus preventing SCA occurrence.

II. Hardware Archtecture

A simplified schematic diagram of the proposed wireless dry-contact monitoring system is shown in Figure 1. It consists of two main parts; an analog front-end that senses and amplifies the cardiac biopotential and a digital part that samples and transmits it to a computer or smart phone for storage and processing. The overall electrode design is based on the works previously presented by Chi et al. [2-4]. Each electrode consists of a small round standard Printed Circuit Board (PCB) with a 26mm diameter which acts as a physical substrate (Figure 2). The biopotentials are sensed through a 250 mm^2 metal plate consisting of solid copper fill on the PCB’s bottom layer covered with a proprietary material. Thus a combination of high impedance resistive and capacitive coupling to the human body is achieved. The front-end amplifier circuit is built on the PCB’s top layer with the precision CMOS operational amplifier LMP7721 (A1, A2) which is selected mainly for its ultra low input bias current (3fA). Special attention is paid to circuit layout and assembly by applying guarding techniques to reduce parasitic leakage current by isolating the LMP7721’s input pins from large voltage gradients across the PC board. The active shielding from external electric field pickup is implemented with an outer ring around the sensing plate and with an inner PCB’s plane just above the sensing plate and the outer ring.

The front-end op-amps (A1, A2) are configured as unity-gain voltage buffers and provide only a signal conversion and no gain. The active shield is driven through R2 by a buffered version of the input signal. This solution has been shown to be effective [4] in guarding the amplifier input without introducing additional loading to the input. To remove the DC offset as well as low frequency noise, a passive RC high pass filter (R3, C1), with a corner frequency of 1Hz, is used. The filter is followed by a low power, precision instrumentation amplifier (A3), which amplifies the difference between the local biopotentials sensed by Electrode 1 and 2, while suppressing the common mode voltage Vcm. The last is implemented with INA333 selected for its micro-power (50μA), zero-drift and rail-to-rail output, thus achieving for low part count. The mid-band differential mode gain is set to 56dB by using 150 ohm resistor value for R7. The connection between the signal ground and the subject, also known as subject grounding is implemented by an actively driven dummy ground electrode (Electrode 3), in order to reduce the common-mode interference. The ground electrode is similar to Electrode 1, 2 but without any electronic components. The common mode signal, Vcm is buffered (A6) and then connected to an inverting amplifier (A7) with a gain of about 50dB for 50Hz. The op-amp output is then fed back to the subject’s body through resistor R9 and the

MOBIHEALTH 2014, November 03-05, Athens, Greece

DOI 10.4108/icst.mobihealth.2014.257236

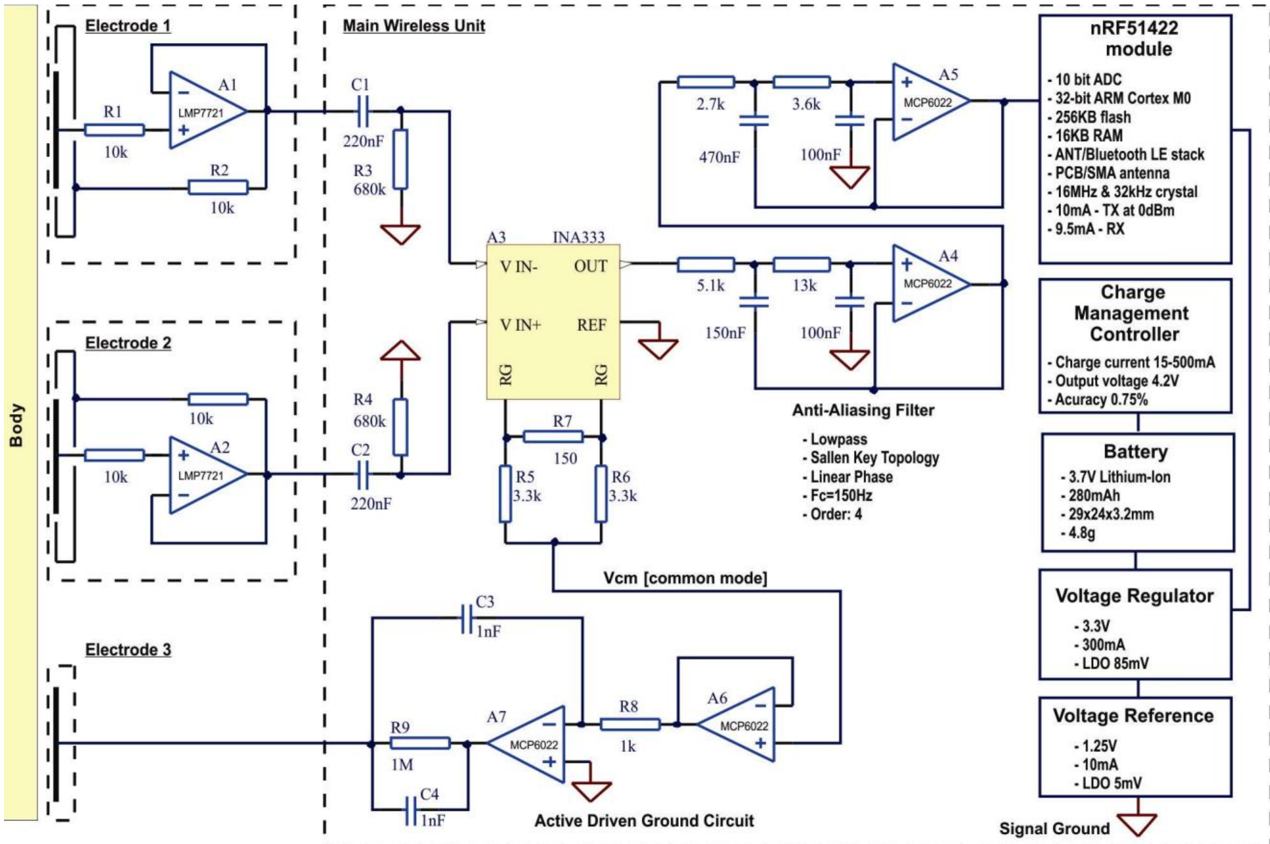

Figure 1. Simplified schematic of the wireless ECG system with dry-contact electrodes.

ground electrode. This circuit theoretically provides a 50dB reduction of common mode voltage, but practically a reduction between 10 and 50 dB is usually accomplished. The signal at the output of A3 is filtered by 4th order 150 Hz low pass filter (A4, A5), to prevent aliasing. Linear phase filter type in a Salen-Key topology is preferred for its excellent transient response. The filtered signals are digitized with an embedded 10-bit ADC.

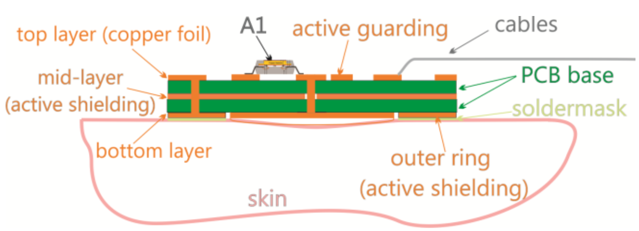

Figure 2. Simplified diagram of the dry-contact electrode.

The main wireless unit contains also a microcontroller and 2.4GHz radio. The last two are implemented with an nRF51422 module [5], based on the ultra-low power system-on-chip (SoC) nRF51422 built around a 32-bit ARM Cortex M0 CPU, thus avoiding the use of an extra microcontroller, reducing the part count and keeping the system dimensions small. The SoC embedded 2.4GHz transceiver supports both ANT and Bluetooth low energy protocols. The system is powered by a 3.7V, 280mAh rechargeable lithium-ion polymer battery providing about 24 hours of continuous monitoring, before recharging. The transmitted signal is acquired by readily available ANTUSB-m stick [6]. It features a miniature design (18.96 x 12.48 x 5.02 mm) that allows users to always leave the stick plugged in.

III. Software Design

A. *Firmware design*

The system firmware handles the periodic sampling of the ECG signal and battery levels utilizing the embedded ADC module, and the communication with a client device using the ANT communications technology. The ECG signal is sampled at 500Hz, while the remaining battery charge level is sampled at 1Hz. The exact sampling times are obtained by enabling the timer/counter module of the nRF51422 SoC, thus generating high priority interrupts every 2ms. The timer module is set to operate at 16bit “timer”mode using a 16Mhz crystal oscillator (HFClk) with a prescaler value of 8, thus operating at 62.5 kHz. The ADC module is set to operate with a sample resolution of 10bit, internal reference voltage of 1.2 Volt and an one-third (1/3) prescaler for the analogue inputs. The results of the signal sampling are stored in one out of four designated buffers using a custom compression scheme in order to achieve optimal use of the buffer space rather than waste bits in padding the 10bit samples into 16bit short values. Whenever such a buffer is filled, the process following a cycle pattern moves on to the next available buffer. The filled buffers are marked as “ready-to-be-sent” via the communication algorithm, and following the successful transmission of their data, the buffers return to an “available for sample storage” state. The buffer size for the ECG signal is 96 bytes, including a header of 5 bytes and a bookend 8 byte packet that signifies the end of a burst data transmission. Using our compression scheme, each such buffer can carry 68 ECG lead values. Given the requirement for sampling rate of 500Hz, it is needed at most eight buffer transmissions per second, which can cover a sum of 68 * 8 = 544 samples. With respect to the communication protocol, the nRF51422 SoC, features a pre-programmed software library (SoftDevice S210) that implements a full ANT protocol stack with a few custom extensions. Our firmware interfaces with this library, while also providing the handler methods for useful events and extensions to ensure reliable transmission of data and fair use of the communication channel amongst all types of data that need to be sent. For our communication purposes, we setup a single ANT channel using the default public Network Number (0), and assigning an RF frequency of 2469MHz. The channel in this firmware version uses the default public network key, and is initialized as a Bidirectional Master Independent Channel (supports one master and one slave device) with a device type of “2” (an agreed upon device “class” number) and device number of “2” (a arbitrary selected ID). The message rate for the channel is set to be 64Hz, with Burst data transmissions. This essentially means that every 1/64 of a second, the ECG ANT device will be given the opportunity to send a “ready-to-be-sent” buffer of sample data using burst data transmission. However not all of these opportunity time slots will be used to send data, because as stated above only 8 burst transmissions are required for the selected ECG sampling rate. The reason that a larger message rate was chosen for the channel, is to facilitate the ability to send other data packets (e.g. data from other sensor modules like an accelerometer, etc.) or retransmit failed burst data sessions. Burst data transmission mode is selected over the default broadcast mode, because it allows the mass transmission of data packets larger than 8 bytes offering success or failure notifications for the transfer of the batch. Our firmware employs techniques to resend a burst transfer if the host MCU of the system is notified of a transmission failure. Also, a custom header is used in the first packet of a burst data session to provide extra information about the sequence of a packet and the length of the transmission.

B. Application Program Design

On the computer side, all sampled physiological data are displayed and saved to a file on hard disk in real-time. The application program running on PC is written in C++ with toolkit Qt version 5.1 for Windows, using object oriented methodology. The application program contains four main parts. The first part handles the graphic user interface (GUI), where OpenGL was used to accelerate the graphic display. The second program part is the communication with the ANTUSB-m device, based on the provided ANT+ libraries [7]. A special protocol was developed which reads the data from the ANTUSB-m stick and handles the transmissions faults or interrupts. The third program part deals with the ECG digital signal processing and includes two digital filters and a real-time QRS detection algorithm for heart rate calculation and future fast detection of pathological cardiac events. The first filter uses a subtraction procedure [8] which does not affect the signal frequency components around the interfering frequency and almost totally eliminates power-line interference from the signal. The procedure applies digital filtering on linear segments of the signal to remove the interference components. These interference components are stored and further subtracted from the signal wherever non-linear segments are encountered. The second filter is a low pass filter aimed to remove the high frequency noise caused from motion artifacts. The QRS detection algorithm is based upon digital analyses of slope, amplitude, and width [9, 10]. The algorithm automatically adjusts thresholds and parameters periodically to adapt to such ECG changes as QRS morphology and heart rate. The fourth program part is the file handler. The data are stored in files with EDF format which is standard format for medical signals [14]. A special class EDFFileHandler was created to manage these files. The application also supports simultaneous multiple recordings, a feature that allows monitoring of more than one subject at the same time. A similar application was developed for Android using the provided Java libraries [7].

IV. RESULTS

Preliminary test measurements were performed with the proposed ECG monitoring system using a compression vest [12] to ensure optimal electrodes fixation to the body by providing firm thoracic elastic enclosure. The subject was a healthy 42-year-old male. Experiments were performed in a standard electrical engineering lab with electrical equipment. Electrode 1 and 2 were positioned on the upper right (right shoulder) and the lower left side (below the heart apex) of the rib cage, corresponding to a standard Lead II torso placement. The ground electrode was placed on the lower right side of the rib cage. Eight preliminary ECG test were performed; with electrodes placed on top of the skin and over a cotton T-shirt while the subject was standing, walking, jumping and running. Figure 3 shows a three second plot of ECG signal obtained with electrodes placed over a T-shirt underneath the elastic compression vest while the subject was walking.

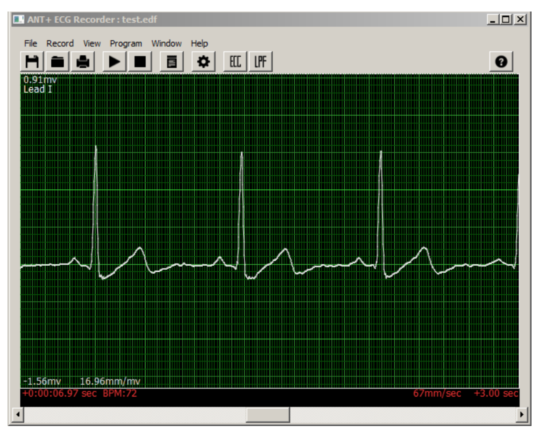

Figure 3. Sample ECG trace measured through a cotton T-shirt over the subject’s chest (Lead II) while walking.

All relevant ECG features are clearly visible and no 50Hz noise is viewed in the signal. As expected, the signal remains mostly undisturbed while the subject is standing or walking. In the cases of running and jumping there were some motion artifacts but the relevant ECG features are still clearly visible and the signal baseline is mostly stable. The system power current consumption was 11mA, measured at +4dBm (2.5 mW) power output and on-air data rate of 250kbps. The maximum measured data transfer range was 65 meters obtained in a building corridor with the same power output, 1/4 Wave Whip Antenna SMA [5] and ANTUSB-m stick [6] as receiver.

V. CONCLUSION

We present the design of a wireless ECG monitoring system using dry-contact electrodes and ultra-low power ANT+ wireless technology. It was shown that by combining an innovative electrode construction, proper PCB, circuit and system design, it is possible to obtain signals on top of the skin and through clothing or hair, suitable for medical-grade ECG applications. The preliminary results showed that the proposed dry-contact electrode can tolerate coupling impedance up to several hundred mega-ohms relatively insusceptible to its variation thus allowing sensing of local biopotentials through clothing or hair. The combination of low profile dry-contact sensors, active grounding, fully isolated battery power and a wireless data transfer, resulted in excellent interference rejection and motion artifact reduction, making the proposed system a potential solution for future mobile applications for health and sports monitoring. However further studies are needed to fully evaluate the proposed system and the influence of skin perspiration, fabric and electrode materials and electrode-skin separation distance over the signal quality. As future work, we plan to add an accelerometer, gyroscope, non-contact thermometer and wearable breathing sensor to the system.

ACKNOWLEDGMENT

This work is supported by BaziFIT Inc. through joint development agreement between Biomedical Technology Unit (BITU), University of Patras, Greece and BaziFIT Inc., USA.

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