Textile electrodes in a body belt to capture and process ECG signals

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Abstract

Life-threatening cardiovascular diseases require early detection or diagnosis. A standard procedure, long-term ECG monitoring of cardiac patients is currently the best way to reduce the number of heart failures. Dry and washable textile electrodes embedded in comfortable garment or in a wearable chest belt have been proven very effective for a long-term ECG monitoring in comparison to the conventional Ag/AgCl electrodes. Hereby, we present a wearable ECG chest belt namely "Active Belt" which contains the stitched textile electrodes for ECG detection and analog preprocessing circuits embedded in tiny cell-phone plugs. The Active Belt performs ECG processing, display and transmission using dual-core OMAP3-based embedded system. Our experiments have shown promising results of textile electrodes along with our hardware and embedded system with a better ECG signal quality for clinical use and hence enabled long-term ECG recording in a daily life.

Keywords: *textile electrode, skin-electrode impedance, long-term ECG, OMAP3, BeagleBoard*

Introduction

Cardiovascular diseases are the main cause of death in the middle-to-old aged people globally. According to WHO estimates, 16.7 million people around the globe die of cardiovascular diseases each year [4]. Early detection of symptoms in the risk group (patients with recent bypass surgery or angioplasty) can significantly reduce the number of heart failures or sudden deaths [1, 2, 3].

Long-term ECG recording is a standard procedure in monitoring cardiac patients. In case of ECG recoding during daily life activity, washable and comfortable textile electrodes can be woven at a precise target location in the wearable garments and hence make the ECG recording less hectic concerning electrode handling and adjustments on the body unlike conventional gelled electrodes.

Over the past few years, a number of wearable physiological monitoring systems have been developed for health monitoring of patients in hospital and real life situations [5]. Some prerequisites for such wearable systems are small size, low-power consumption, low weight, real-time signal processing and most importantly wireless connectivity [6]. Processors like the OMAP family from Texas instruments have a dual-core functionality (ARM+DSP) and are suitable for wearable health monitoring applications [7].

Concept

Hereby, we present an Active Belt which is a wearable long-term ECG recording system developed in our laboratory (Fig. 1). Textile electrodes are stitched on the skin-contact side of the belt. The electrodes are wired to a small cell-phone connector, which contains a biopotential amplifier for signal amplification and filtering. The plug sends conditioned signals to an ADC plug for digitization. Consecutively, we process, display and wirelessly transmit the digitized ECG signals by utilizing the power of a dual-core OMAP3-based embedded system.

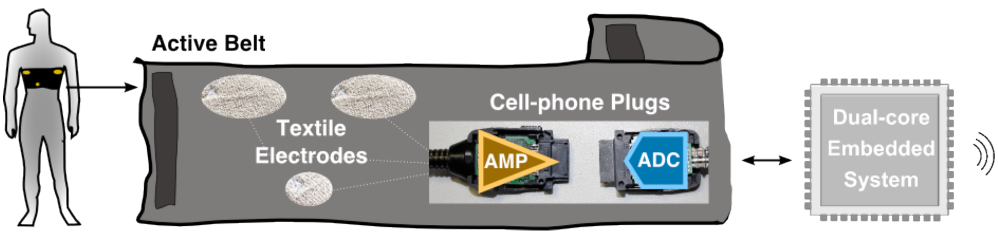

Fig. 1: The Active Belt concept

Materials & Methods

Active Belt & Textile Electrodes

A stretchable and breathable chest belt (34.5cm x 84.5cm) was fabricated from neoprene material (SEDO Chemicals Neoprene GmbH, Germany) with velcro on the closing ends. Three Textile electrodes, two of them elliptical (⋯6.5cm and ⋮3.5cm) for ECG were stitched on two sides of the torso on the belt and a circular electrode for ground (Ø2.5cm) next to the navel as shown in Fig 2 b.

The washable textile electrodes (TITV Greiz, Germany) are made of Polyamide threads (ELITEX^R) coated with pure silver with a thickness of 1-2µm and a resistivity of 20Ω/m [8]. Each thread can be stretched up to 7% of the original length without compromising the conductivity [8].

ECG Analog Circuits in Cell-phone Plugs

As shown in Fig. 1, the ECG analog circuits are embedded inside two 1.6cm x 2cm cell-phone plugs (AXR72161, Masushita Electric Works, Ltd). Amplifier and ADC plugs are thus mounted on the Active Belt next to the electrodes.

The raw ECG signal detected by the textile electrodes on the body is fed to the amplifier plug, which contains the bio-potential amplifier array RHA1016 with additional components. The low-power RHA1016 contains 16-channel differential amplifiers (gain = 200) with built-in programmable 3rd order butterworth low pass filter . The value of the external resistors R_1 and R_2 sets the lower cut-off frequency of 200Hz.

The amplifier's output connects to the ADC plug, which contains a single channel ADC (ADS1271, Texas Instruments Inc., USA) with a data rate up to 105kSps following an instrumentation amplifier (IA) with a gain of 20. The ADC is controlled by an industry standard serial peripheral interface (SPI) from an external source, which in our case is the dual-core embedded system.

Dual-core OMAP3-based Embedded System

OMAP3530 (OMAP3) (Texas Instruments Inc., USA) found in new generation smart-phones is a dual-core (ARM+DSP) application processor, which is an ARM-based SoC with laptop-like performance. The OMAP3 has the following prominent features [10] (see Fig 2 c):

- 720 MHz ARM Cortex-A8 core provides 1400 Dhrystone million instructions per second (MIPS) and runs fully-featured operating systems e.g., embedded Linux, WindowsCE, Android.
- 520 MHz C64x+™ DSP furnishes excellent signal processing and computing.
- specialized subsystems provide display connectivity.
- power and clock-management scheme enable high-performance and low-power consumption.

The USB-powered BeagleBoard is an OMAP3530 based development board and features all needed components in order to access the peripherals such as mouse, keyboard, monitor, speakers, mic, memory cards/sticks, internet, ADC, FPGA etc [11]. We chose a user-friendly linux distribution "Ångström" for the BeagleBoard [9]. The Ångström is a distribution with a highly portable and reconfigurable core, built using the OpenEmbedded build system with special attention to embedded devices [9]. The BeagleBoard utilizes multi-channel serial port interface (McSPI) and hence communicates with the ADC via SPI. In case of multi-channel ECG recordings, GPIO on the BeagleBoard controls the channel selection for he RHA1016.

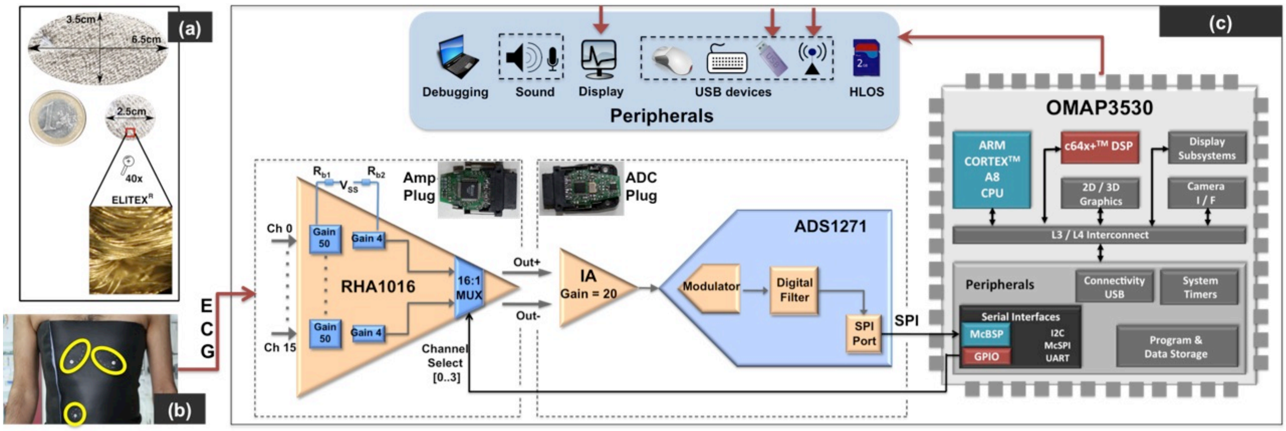

Figure 2: (a) The textile electrodes and their dimensions and (b) the textile electrodes (marked yellow) stitched on the Active Belt. (c) A raw ECG signal from the textile electrodes is amplified, filtered and digitized in the cell-phone plugs containing analog circuits, the RHA106 and the ADC. The digital data is fed to the OMAP3 via SPI port. The OMAP3-based embedded system facilitates on-site visualization and storage along with wireless data transfer to a remote computer

Results & Discussions

All experiments presented in this report are conducted on a male subject of age 35.

Textile Electrodes under Test for Long-term Recordings

In long-term surface biopotential recordings, the skin-electrode impedance plays a major role in the quality of signals. The body impedance depends on several factors such as size and location of the contact electrode as well as the skin hydration. Experiments were conducted to measure the skin-electrode impedance of the textile electrodes in comparison with commercial gelled disposable ECG electrodes with the help of an USB amplifier (g.USBamp, Guger Technologies, Austria).

As shown in Fig. 3 a, the conventional electrode has a flat response over time. The impedance measured from the textile electrodes on the Active Belt shows exponential decrease over time. Our hypothesis behind such exponential decline in impedance is that the sweat secreted by skin under the textile electrode location enhances the contact. To further investigate our theory, we prepared a sweat phantom (saline-0.9% NaCl) and injected it into one of the textile electrodes. At the same moment, we observed a plummeting of the contact impedance and afterwards its gradual rise again as the saline evaporated.

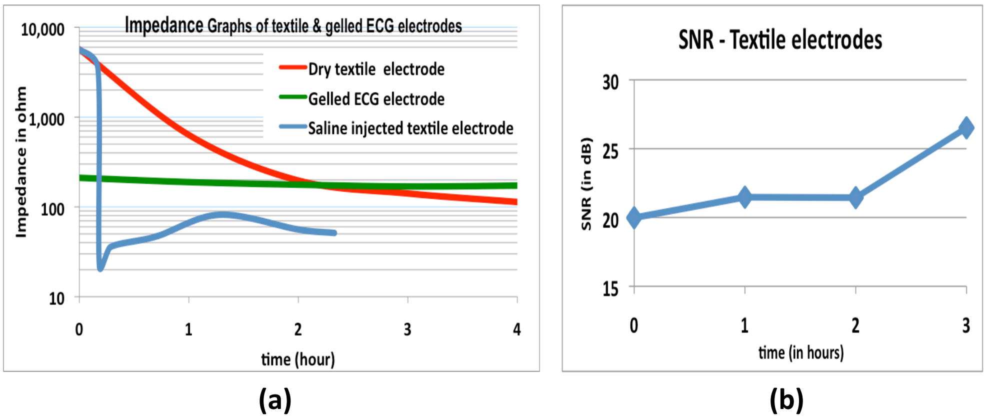

Figure 3: (a) Impedance graph of the textile and conventional ECG electrodes and (b) SNR over time for the textile electrodes

Another experiment was conducted to see the performance of the textile electrodes for long-term ECG recording. Signal-to-noise ratio (SNR) was calculated for the ECG signals captured by the textile electrodes with the USB amplifier at intervals of 1 hour. Fig. 3 b depicts an SNR rise as the textile electrodes improve their skin contact and hence enhance ECG signal quality. This corroborates that the skin-electrode impedance for textile electrode decreases in a long run.

The featured results from the textile electrode experiments reveal that the textile electrode is an appropriate candidate for the long-term ECG recording although it requires further research on the artifact removal.

Active Belt under Test for ECG Recordings

Fig. 4 a shows the ECG signals captured by textile electrodes as well as by the conventional Ag/AgCl ECG recording in connection with commercial monitors. It is apparent that the textile electrode possess a good quality in terms of ECG detection.

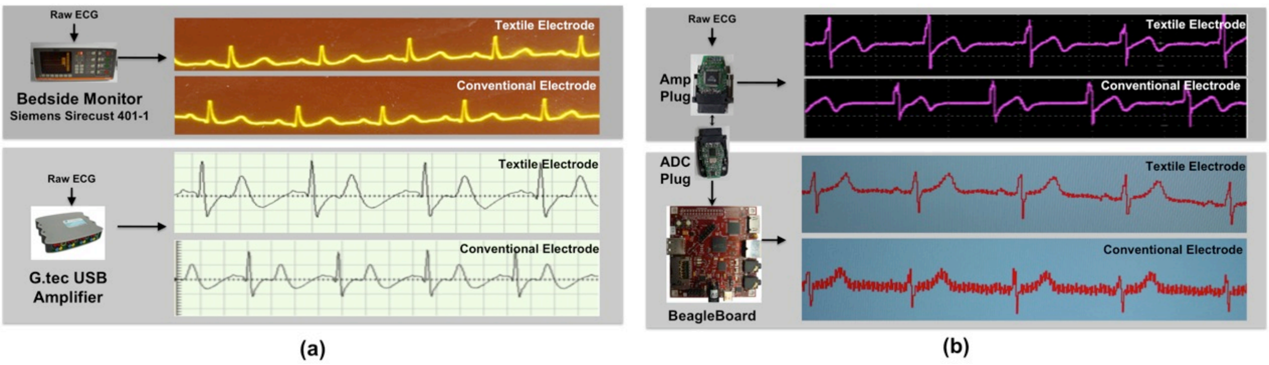

Figure 4: (a) The screenshots of the ECG recorded by the commercial monitors and (b) the screenshots of the ECG recorded by the amplifier plug and the complete Active Belt system

Fig. 4 b depicts the ECG signals as an output of the amplifier plug. These amplified signals are indistinguishable from the signals of the commercial amplifiers and hence verify the performance of the amplifier plug circuit. The processed ECG signals arriving at BeagleBoard represented in Fig. 4 b shows that the ECG signals from convention electrodes are obviously contaminated by power-line noise more than the textile electrode signals. It has been observed that the ECG signals out of the ADC plug have very high resolution (24-bit) at sampling frequency of 105 kSPS. Our C++ embedded application running on the ARM side of OMAP3 fetches the digital ECG signals from the ADC and converts them into decimal values. The application also plots the signals in real-time on the Ångström desktop environment utilizing SDL library.

Conclusions & Outlook

We have presented a successful implementation of a wearable single channel ECG Active Belt for a long-term ECG recording. Initially, the textile electrode was compared with the conventional electrode in terms of their ability to perform long-term recording. The textile electrode has shown a promising outcome by exhibiting a low skin-electrode impedance and a high SNR for long-term recording. We have brought the analog processing hardware in tiny cell-phone connectors which are detachable and mounted in vicinity to the ECG detection site. We utilized the computing power of the dual-core processor OMAP3 for ECG processing, display and transmission. The OMAP3-based BeagleBoard receives high resolution ECG digital signal and facilitates the real-time display. The wireless transmission of the ECG signals from the Active Belt via WLAN is being realized and will enable the remote diagnosis of the wearer (patient). Currently, the efforts are being made to design sophisticated battery-based power supply for the Active Belt. At the end, the miniaturized OMAP3-based custom board with minimum components will complete the working Active Belt that will bring a long-term ECG recording in patient's daily life and help the experts to diagnose cardiovascular problem earlier.

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Note: The featured work is a part of a doctorate thesis on "**Multimodal Biosignal Monitoring based Dual-core Embedded Platform".**