Sensors & Transducers, Vol. 183, Issue 12, December 2014, pp. 42-47

Sensors & Transducers

Design and Integration of Wearable Devices in Textiles

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Received: 15 October 2014 /Accepted: 28 November 2014 /Published: 31 December 2014

Abstract: In this article, the design, production method, integration and characterization of textile sensors for the continuous monitoring of cardiac and respiration vital signals are presented. Textile electrodes, capacitive and piezoresistive sensors and respective interconnect plate were developed and integrated in elastic and adjustable chest bands, using a 6-needle digital embroidery machine and electrically conductive commercial threads. The signal´s waveforms were recorded via PC with a data acquisition module and a LabView program. The signal to noise ratio of textile electrodes, having distinctive surface morphologies, that were either textured or smooth accordingly with the embroidery pattern used, were analyzed with Matlab. The quantitative method indicated differences between the two types of textile electrodes but performances comparable to standard Ag/AgCl gel electrodes. The sensors and interconnect plate were fully realized with the embroidery stitching method with textile fabrics and threads, and have a compact design, are lightweight and washable. The method offers great versatility for custom demand, in terms of sensor design and materials. *Copyright © 2014 IFSA Publishing, S. L.*

Keywords: Textile electrodes, Embroidery technique, Vital signals, Piezoresistive sensors, Capacitive sensors, Integration, Wearable electronics, Telemedicine.

1. Introduction

Wearable electronics combining low-power circuits, wireless transmission and advanced packaging have allowed important breakthroughs in medical applications [1] and smart clothing [2-3] with functionalities to monitor and record vital signals in ubiquitous and non-invasive ways. These were considered strategic to lower morbidity and health care costs associated to diseases of the circulatory system. Holter monitors (ambulatory electrocardiography devices) are currently used to capture a patient electrocardiogram (EKG) for one to three days, using standard gel electrodes taped to patients' skin. The EKG data is analyzed afterwards by doctors to evaluate the patient´s cardiovascular condition. Although Holter systems are portable, the patients feel the cables connecting the electrodes to the electronic unit as intrusive and uncomfortable. Wearable technologies to monitor vital signals had a strong impulse around 2000 with public and private funded projects that produced important outcomes, such as the French VTAMN [4], the EU Wealthy [5] and MyHeart [6] and the American LifeShirt [7]. The Wealthy system consists of elastic garments made by seamless knitting technology, integrating knitted electrodes and piezoresistive sensors to capture EKG and pulmonary ventilation signals, respectively [8].

The electrodes and interconnects were knitted with electrically conductive yarns, consisting of stainless steel surrounded by twisted cables of viscose. Hydrogel membranes placed between the knitted electrodes and the skin were necessary to improve signal to noise ratio (SNR) and minimize skin irritation [9]. The piezoresistive sensors were realized with fabrics of lycra coated with carbon-loaded rubber and a commercial electroconductive yarn. A handheld electronic unit was used for signal conditioning, processing, storage and wireless GPRS cellular communication. The LifeShirt used gel electrodes and embedded textile sensors for plethysmographic respiration monitoring and a small personal controller held in a pocket to capture signals and transmit data via Bluetooth [10]. The VTAM system also consisted of a T-shirt integrating smooth, dry ECG electrodes, with leads and treatment modules incorporated into textile woven fabrics. It used a motherboard, transmission module and power supply mounted on a belt that was connected to the VTAM T-shirt through a microconnector [11]. Screen printed silver paste has also been proposed to produce planar fashionable circuit board electrodes and to perform wire bonding of electronic chips on fabric [12]. A flipChip on flex intermediate connection layer, having higher density than textiles can provide, was proposed to interconnect high density silicon chips to textile fabric [13]. More recently, an active belt integrating washable textile electrodes from TITV Greiz, and two cell-phone plugs (1.6 cm × 2 cm) for signal processing was proposed [14]. Nowadays Holter systems have very small dimensions, but connecting to Ag/AgCl gel electrodes through long cables hanging on the chest of the user, and efforts continue to improve the performance of textile sensors and interconnects, and the electronic systems integration in clothing, involving the interface electronics – textiles. A question often put forward concerns the main challenges to manufacture smart textiles. Large and small companies debating these questions, point out the difficulty to combine Textile and Electronic Manufacture processes. Moreover, there is a dilemma concerning where to put the interface on the technical level in the supply chain between the electronic manufacturing service and the textile manufacturing service [15].

Another challenge concerns availability of human resources with competences to deal with the interface textile / electronics manufacturing.

The embroidery technique has been proposed as a technique with great potential and versatility for smart textiles [16].

In this work we present an embroidery stitching method [17] that is suitable for large scale production, and that offers versatility in terms of product design and materials for custom demand. Textile electrodes, having distinctive surface textures and electrical characteristics provided by embroidery patterns and electrically conductive threads were produced and characterized in terms of SNR. Textile capacitive and piezoresistive sensors were also produced with the embroidery method and their response to normal, accelerated and deep breath conditions was measured.

2. Design, Materials and Processes

The EKG sensors were conceived to be replaceable but washable and long-lasting, comfortable to wear in contact to the skin and inexpensive. They consisted of two layers of fabrics having a circular shape with a diameter of 25 mm. One fabric layer had defined a circular embroidery pattern with a diameter of 16 mm, produced with electrically conductive threads; the other fabric layer, corresponding to the back side of the sensor, integrated a snap fastener applied by pressure with a Universal press. Two types of embroidery patterns were produced, to provide the electrodes with two distinctive surfaces morphologies, one smooth the other textured and address their influence in the electrical contact electrode – skin by measurements of SNR. The sensors integrated snap fasteners for attachment to an interconnect plate in an interchangeable manner, and to obtain a direct comparison of SNR performance between textile and gel electrodes.

An interconnect plate was produced by layering either a semi rigid foam or rubber plate between two fabrics, one of the fabrics having defined interconnect lines terminating at one end in female snap fasteners for the attachment of the EKG sensors, and terminating at the other end, in snap fasteners for connecting the sensors to an electronic data acquisition module. The interconnect lines were produced with a stitching pattern using conductive threads and were protected and electrically insulated with embroidery patterns using threads of polyester.

A piezoresistive sensor and two capacitive sensors were realized for the continuous monitoring of pulmonary ventilation (PV). The piezoresistive sensor consisted of two superimposed sinusoid-like embroidery patterns realized with an electrically conductive elastic thread. An insulator sheet of cotton fabric prevents short circuits between the two patterns, having a length each of about 20 cm and a total resistance of 30 Ohms.

The capacitive textile sensors consisted of a conductive fabric of bamboo with 30 % silver from Less EMF Inc. (USA), having a sheet resistivity of less than 1 Ohm and a rectangular shape with lateral dimensions of 4.5 cm × 5.0 cm.

The textile electrodes and interconnects used electrically conductive threads Silverpam 250 from the French manufacturer Tibetech and silver plated Nylon from Less EMF Inc., USA, having an electrical resistance of 200 Ohm $\cdot$ m^{-1} and 40 000 Ohm $\cdot$ m^{-1}, respectively.

The piezoresistive sensor was made with a piezoresistive elastic thread, Elitex HE, from the German manufacturer Imbut GmbH.

The standard gel electrodes that were used for comparison consisted of Ag/AgCl Cleartrace RTL – 1700C from ConMed, USA.

The EKG and PV sensors and interconnect plate were integrated in elastic and adjustable chest bands made of neoprene and fabrics of elastane – polyimide. Velcro stripes allowed the adjustment of the chest bands to the body of the wearer. In Fig. 1 are shown photos with views of a chest band integrating an interconnect plate with connections for three EKG sensors and one neutral electrode. Photos of EKG textile and standard electrodes are also shown in the figure.

(a)

(b)

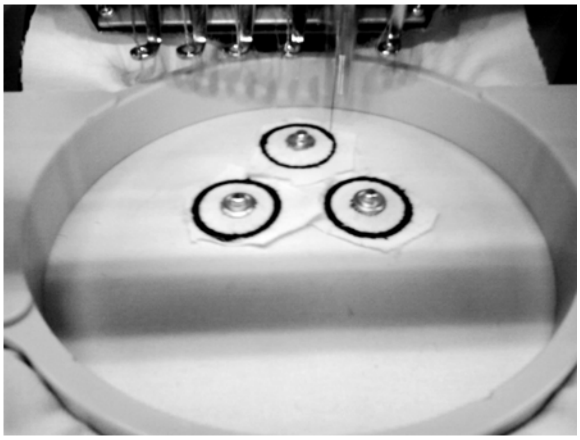
Fig. 1. a) Photo of chest band with interconnect plate having defined interconnections for three EKG sensors and one neutral electrode; b) photo of standard Ag/AgCl electrode and textile electrodes.

In Fig. 2 is shown a view of a chest band integrating thee electrodes and two capacitive sensors for the continuous monitoring of the EKG and PV signals, respectively.

Fig. 2. Photo of chest band with interconnection baseplate of neoprene integrating three textile electrodes and two capacitive sensors for the monitoring of EKG and PV signals, respectively.

The embroidery method used to produce the textile sensors is versatile in terms of design and materials. Furthermore, it allows to fully produce the wearable systems with no intermediate steps in a clothing´s manufacture workshop.

A snapshot view of the production process of the EKG and piezoresistive sensors with a SWF digital embroidery tool is shown in Fig. 3.



(a)

(b)

Fig. 3. Photo with views of production process with 6-needle digital embroidery stitching tool: (a) of textile electrodes; (b) piezoresistive sensor on neoprene.

3. Experimental

An elastic and adjustable chest band made of neoprene and integrating the interconnect baseplate and the piezoresistive sensor was used with a male volunteer, seat motionless in a relaxed position. The interconnect baseplate was positioned on the left side of the thorax, in the region of the fifth intercostal spaces, with the two EKG sensors spaced apart 4 cm from center to center.

Each pair of terminals from the EKG and PV sensors were connected each to the analog inputs of a programmable data acquisition unit, NI-DAQ 6212 with a PC/USB interface, in differential mode. A LabView program, using a band-stop and a band – pass Bessel filters of order 6, of 50 Hz and 0.05 – 100 Hz, respectively, were used. The program graphically display and record in a PC the waveforms.

The piezoresistive sensor had a static resistance of 30 Ohms and was connected to a voltage divider circuit supplied with a 9 V battery. The electrical circuit with the piezoresistive sensor is represented in Fig. 4.

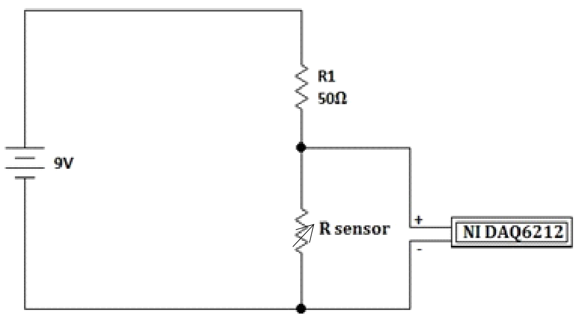


Fig. 4. Electrical circuit with piezoresistive sensor and differential connection to data acquisition module.

The SNR amplitude of the textile and standard Ag/AgCl gel electrodes were obtained with a Matlab program. The program extracted from each experimental data file, 50 waveforms and applied a smooth function, with a span = 150, to remove any baseline distortion or shift and the P and T segments. From the raw and correspondent smooth data matrixes, the program calculated for each k experiment of a specific type of electrode, the mean values of signal amplitude, *Vs, k* of the QRS complex, and of noise amplitude, *Vn, k,* based on 100 points of a segment away from the QRS complex, described by the following expressions:

$$V^{s,k} = \frac{1}{50}\sum_{i=1}^{50}\left(\left|V_{max}^{s,k,i} - V_{min}^{s,k,i}\right|\right), \tag{1}$$

$$V^{n,k} = \frac{1}{50}\sum_{i=1}^{50}\left(\frac{1}{100}\sum_{j=1}^{100} 2\cdot\left|V^{n,k,j}\right|\right) \tag{2}$$

The SNR is given by the ratio of the mean values of signal to noise,

$$SNR^{k} = 20\,log\,\frac{V^{s,k}}{V^{n,k}} \tag{3}$$

For each type of textile electrodes, 20 sequential experiments were performed. Every three experiments each pair of electrodes was replaced with a similar pair, every type of electrode represented by six nominally identical electrodes.

Test structures consisting of stripes having lateral dimensions length x width = 50 mm × 5 mm and 120 mm × 6 mm were made using embroidery patterns identical to those constituting the smooth and textured electrodes. The stripes were terminated in circular pads for electrical contact to two probes to measure electrical resistance. The shorter stripes were used in the washing resistance tests, the longer stripes in the abrasion tests, described next.

Washing resistance tests were performed with a Linitest machine following the guidelines of the ISO 105 of 1978. The tests consisted of machine washing cycles with a detergent solution, at 40 °C and duration of 45 minutes followed by a rinsing step. The samples were dried with an air dryer. Detergent solutions of 600 ml were prepared with DI water and a concentration of 1 g/l of a mild soap. Test structures consisting of stripes had the electrical resistance measured prior to laundering. Textile electrodes and test structures were submitted to laundering cycles and had the SNR amplitude and resistance measured, respectively.

Abrasion resistance tests of the textile electrodes were performed with a Mark II tester from Shirley Developments Ltd and the tests structures previously described. An adaptation of the Martindale method was used. In this adaptation, the places of the sample and abrasive fabric were interchanged, in order to accommodate the long test structures, more suitable to measure the changes of the electrical characteristics of the embroidery patterns as a function of the number of abrasion cycles, N = 5000, 10 000 and 30 000.

4. Results and Discussion

In Fig. 5 is shown the mean SNR and standard deviation (error bars) for the two types of textile EKG sensors, smooth and textured electrodes, and the gel electrodes.

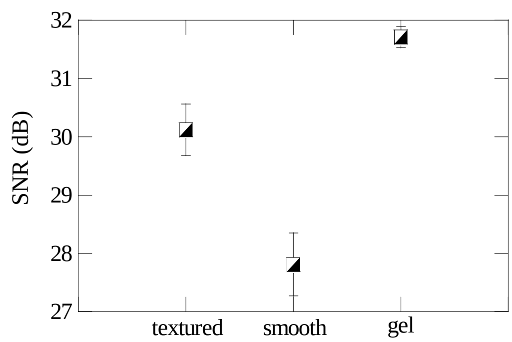


Fig. 5. SNR ratio amplitude of textile and gel electrodes.

The textured electrodes exhibited a SNR performance very similar to that obtained with the gel electrodes, followed closely, 2 Dd below, by the smooth electrodes. The results suggest that the textured surface provide better electrical contact between the electrodes and the skin. The SNR results presented in here are considerably better than those reported by other authors, for both types of textile electrodes. The SNR amplitude obtained with gel electrodes, of 31.5 dB, is similar to those reported by other researchers [12]. The textured electrodes, with a SNR $\cong$ 30 $\pm$ 0.5 dB, present a performance considerably better than those presented in the literature [12], [18]. Furthermore, no significant difference of performance was obtained in either conditions of long acquisition time or under either sweat or moisture of the electrodes. The fact that our textile electrodes show good SNR with no need of moisture is considerably different to the results extensively reported in the literature [18-20].

EKG waveforms of textile and Ag/AgCl gel electrodes and the response of the textile piezoresistive and capacitive sensors to various modes of respiration are shown in Fig. 6.

In agreement with the SNR results, the cardiac signals obtained with the textile electrodes show higher noise than the obtained with standard Ag/AgCl gel electrodes. The EKG waveforms exhibit identical features, the QRS complex and T wave well defined.

The response of the PV sensors exhibit good signal amplitude in the following three modes of breath; normal, rapid and deep. The capacitive sensors may be advantageous in relation to the piezoresistive sensors because they do not require a voltage divider circuit with a battery to supply current and exhibit a signal output about ten times higher than the piezoresistive sensor.

The two types of textile electrodes and the textile piezoresistive and capacitive sensors appear adequate to the continuous monitoring of the cardiac and respiration rhythms.

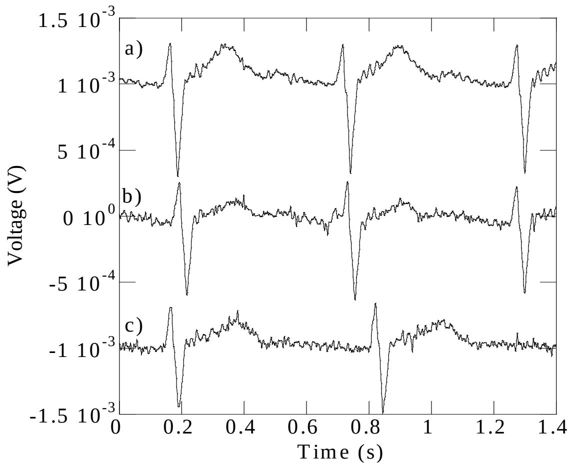
The textile electrodes and rectangular test structures having defined on fabric identical embroidery patterns to those used in the textile electrodes were submitted to 30 washing cycles. The electrical resistance of the test structures was monitored between washing cycles and no significant variation was obtained as shown in Table 1.

In Table 1, R_0 and R_f relate to the electrical resistance of the test structures before washing and after 30 washing cycles, respectively; SNR^0 and SNR^f relate to the signal to noise ratios of textile electrodes before washing and after 30 washing cycles, respectively.

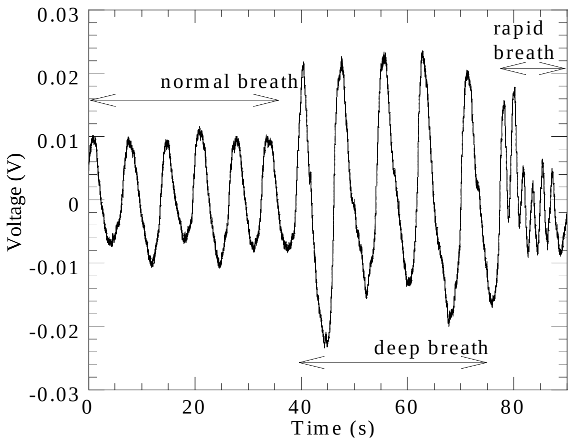
The textile electrodes that were submitted to 30 washing cycles do not show a measurable difference in the SNR amplitude in relation to the un-washed electrodes.

Abrasion tests realized with an adaption of the Martindale method and using up to 20 000 abrasion cycles show no changes in the electrical resistance of the test structures.

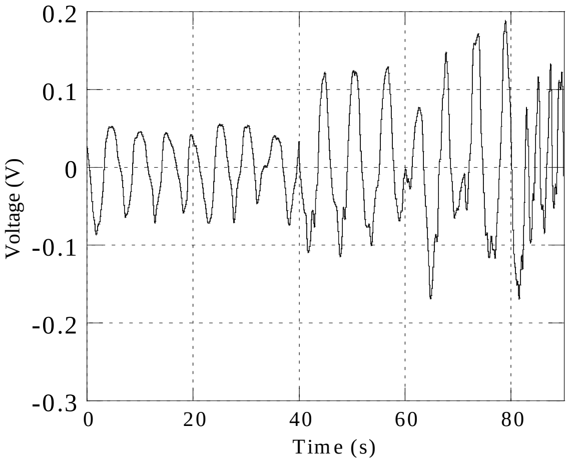
The experimental results obtained with washing resistance and abrasion tests indicate that the sensors have very good resistance to wear, provided by the high compactness of the embroidery patterns used, as the electrically conductive threads are relatively fragile. The sensors can be washed and re-used several times until their performance starts to degrade.



(a)



(b)



(c)

Fig. 6. a) EKG waveforms obtained with gel and textiles sensors, where (a) gel electrodes, (b) textured electrodes, (c) smooth electrodes; b) piezoresistive sensor response to normal, deep and rapid breath rates; (c) capacitive sensors response.

Table 1. *R* and *SNR* variation with 30 washing cycles.

Type	R_0 (Ohm)	R_f (Ohm)	SNR^0 (dB)	SNR^f (dB)
Textured	0.74	0.93	30.1 ± 0.4	29.4 ± 0.1
Smooth	1.14	1.43	27.8 ± 0.5	NA

5. Conclusions

The embroidery method presented shows great degree of versatility to produce textile sensors and interconnects and for their integration in clothing systems. Moreover, the method is suitable for large scale production and custom demand because is based on a digital system.

The SNR amplitude obtained with textile electrodes having distinctive surface morphologies confirm that the electrode's performance is dominated by the skin-electrode electrical contact and the surface of the electrode is an important aspect of the sensor design. The embroidery method is suitable to tail the electrical and surface characteristics of textile electrodes with adequate choice of conductive threads and embroidery patterns.

The method is also suitable to produce and integrate into clothing other types of textile sensors and electronic components, such as light emission diodes (LEDs), antennas, batteries, etc.

Acknowledgements

This work was financed by FEDER funds, through the Operational Program Factors of Competitiveness – COMPETE and by National Funds through FCT – Science and Technology Foundation, under the grant PTDC/EEI-ELC/1838/2012.

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