Performance Evaluation of Sensing Fabrics for Monitoring Physiological and Biomechanical Variables

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Abstract—**In the last few years, the smart textile area has become increasingly widespread, leading to developments in new wearable sensing systems. Truly wearable instrumented garments capable of recording behavioral and vital signals are crucial for several fields of application. Here we report on results of a careful characterization of the performance of innovative fabric sensors and electrodes able to acquire vital biomechanical and physiological signals, respectively. The sensing function of the fabric sensors relies upon newly developed strain sensors, based on rubber-carbon-coated threads, and mainly depends on the weaving topology, and the composition and deposition process of the conducting rubber-carbon mixture. Fabric sensors are used to acquire the respitrace (RT) and movement sensors (MS). Sensing features of electrodes, instead rely upon metal-based conductive threads, which are instrumental in detecting bioelectrical signals, such as electrocardiogram (ECG) and electromyogram (EMG). Fabric sensors have been tested during some specific tasks of breathing and movement activity, and results have been compared with the responses of a commercial piezoelectric sensor and an electrogoniometer, respectively. The performance of fabric electrodes has been investigated and compared with standard clinical electrodes.**

Index Terms—**Biomechanics, sensing fabric, vital signs.**

I. Introduction

An emerging concept of health care, based on continuously monitoring vital and behavioral signs to provide assistance to patients and health care users, is gaining wide consensus [1]–[4]. Advances in sensor technology, as well as in communication technology and treatment of data, have stimulated the creation of a new generation of health care monitoring and diagnostic systems [5]–[7]. The use of functional materials enables design and production of a new generation of garments with distributed sensors and electrodes [8]–[10], allowing the noninvasive monitoring of health status. Fibers and yarns with specialized electrophysical properties can be integrated and used as basic elements to fabricate woven or knitted functional textile systems [11], [12]. These integrated systems enable the detection of electrical [electrocardiogram (ECG), electromyogram (EMG)] and mechanical (respirogram, motor activity) physiological signals [13], [14]. Wearable comfortable systems aid the daily acquisition and processing of multiparametric health data, providing an early detection of pathological signs and improving the curative rate of disease without interfering with daily life [15], [16]. This paper is aimed at assessing the properties of fabrics able to acquire bioelectrical and biomechanical signals, in particular, the sensing properties of bioelectrical fabrics sensors, and the biomechanical performance related to the piezoresistive properties of fabric sensors. In order to confer special functions to fabric, the intrinsic properties of materials have been modified by means of dedicated treatments. Bioelectrical (ECG and EMG) and biomechanical (respirogram and movement activity) signals acquired through sensing fabrics are compared with signals acquired through standard electrodes and sensors commonly employed in clinical and experimental studies. The comparison had been done both in terms of engineering assessment and clinical validation, i.e., each signal has been quantitatively evaluated in the time and frequency domain to make sure that no medical and diagnostic information is lost. Moreover, it is worth emphasizing that although fabric sensors and electrodes should be thought to be integrated in a garment, the aim of this paper is to investigate the performance of the single fabric electrode and sensor, without envisioning their behavior in the context of a sensorized shirt. In order to validate the performance of fabric sensors and electrodes, their response in the time and frequency domains has been compared with the outputs of commercial devices. Concerning the ECG, surface EMG, and respirogram, the comparison has been performed by extracting the most significant parameters at rest conditions. Due to motionless conditions, the acquired signals are quite artifact-free, and the noise level is partially suppressed, not least because it is suitably filtered. Further studies are currently being performed in order to analyze recordings while subjects are moving.

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II. Materials and Methods

A. Fabric Electrodes

Conductive fabric electrodes have been realized with commercial stainless steel threads twisted around a standard continuous viscose textile yarn. They were woven using the tubular intarsia technique to get a double face, where the external part

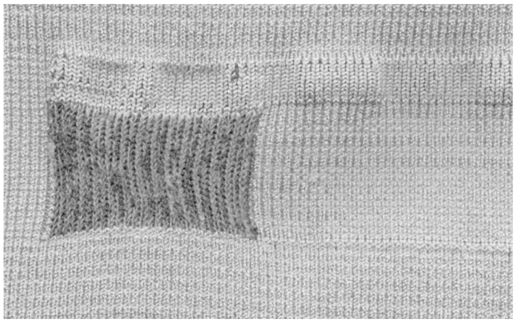

Fig. 1. Enlarged view of the structure of a fabric electrode for ECG and EMG recordings integrated into a sensorized shirt. The darker area in the pattern is yarned with conductive threads, while the surrounding lighter region is realized with the nonconductive basal yarn acting as insulator.

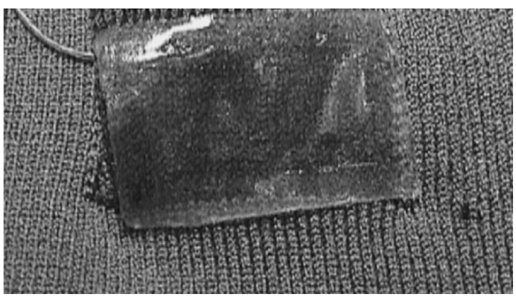

Fig. 2. Hydrogel membrane used to improve the skin-electrode coupling. It covers the conductive area coming into contact with the skin.

is realized with the basal yarn, not sensitive, to isolate the electrode from the external environment (see Fig. 1). To improve the quality of the electrical signal in dynamic conditions, a hydrogel membrane, obtained from ST&D Ltd (Belfast, U.K.), has been used (see Fig. 2). The membrane reduces the contact resistance between the skin and the electrode, as well as improving comfort. In addition to the presence of a bactericide/fungicide, the hydrogel membrane has been chosen with a pH range between 3.5 and 9, hence causing minimum skin irritation. The real challenge would be the removal of the membrane, allowing a long-term electrode entirely made of fabric to be available, but, in this case, the crucial issue of the electrical coupling between electrode and skin should be suitably addressed. Conductive yarns and coated fabrics are resistant to repeated washing in aqueous solutions, without decreasing their performance.

B. Acquisition of Bioelectrical Signals Through Electrodes

The ECG and EMG electrodes based on conductive fabric have been tested in comparison with standard ECG clinical electrodes (RedDot by 3M) and standard surface EMG Ag/AgCl electrodes, respectively. Preliminarily, the electrical impedance in the transversal direction of fabric electrodes has been measured across two plates, having the same surface as the electrodes, and covered by adhesive copper, pressed on both electrode's faces.

Since both ECG and EMG are low-intensity signals, they have been suitably amplified and filtered by means of a GRASS-TELEFACTOR model 15LT device, equipped with differential amplifiers, model 15A54 and gain 1000. Next, signals were acquired by a National Instruments acquisition card PC-6036E.

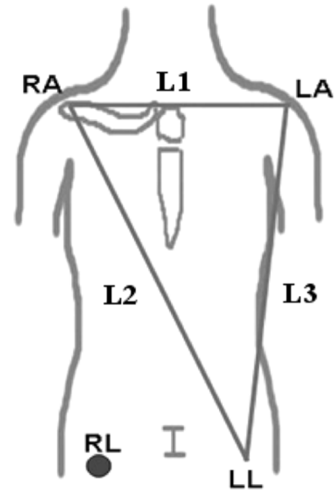

Fig. 3. ECG leads located according standard configuration of Einthoven. LA and RA: left and right arm, respectively. LL and RL: left and right leg, respectively. L1, L2, and L3: bipolar Einthoven leads.

1) Electrocardiogram (ECG): The electrodes (Red Dot and fabric) were positioned according to standard Einthoven configuration, with two electrodes placed onto each root arm and one electrode onto left root leg. The bipolar lead L1, which is the upper side of the conventional Einthoven's triangle configuration, and measures the potential difference between the two arms, has been selected for our experiments. The right-leg electrode (RL) has been also used as reference (see Fig. 3). Besides being largely used for clinical investigation, this lead is also quite immune to artifacts due to small movements of the torso. The GRASS-TELEFACTOR was set with the bandpass filter between 1 and 100 Hz, notch filter at 50 Hz. Moreover, the National Instrument acquisition card was set with a sampling rate of 1000 Hz. The filter settings were chosen in order to preserve the most significant frequency content of the ECG signal within the bandwidth usually analyzed in diagnostic and monitoring activity, according to the American Heart Association (AHA) recommendations (0.1–100 Hz) [17]. The high sampling frequency allows postacquisition digital elaboration and noise filtering.

The evaluation of performance of fabric electrodes with respect to standard clinical RedDot electrodes has been done, comparing the most significant time parameters extracted from the morphology waveform (P-waves, QRS complexes, ST segments, and T waves), and performing a spectral analysis [fast Fourier transform (FFT)] on signals coming from both electrodes. In addition, the magnitude squared coherence function, estimated by using the toolbox COHERE (Matlab Program), was calculated on ECG signals acquired with both fabric and Red Dot electrodes. The coherence is expressed by the ratio between the power spectra of two signals and the cross spectrum of same signals

$$C_{xy}(f) = \frac{|P_{xy}f|^2}{P_{xx}(f)P_{yy}(f)}. \quad (1)$$

Since fabric electrodes are made of standard textile filaments twisted together with thin stainless-steel threads, we tackled the crucial issue of the polarization phenomena. In the literature, it

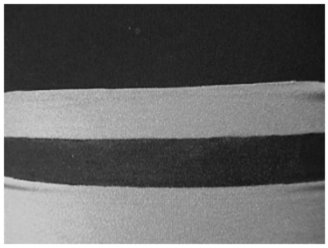

Fig. 4. Strain fabric sensor, basically a textile band wrapping around the chest, used for monitoring both respiration and movement signals.

is reported that stainless-steel electrodes are far more polarizable than Ag-AgCl electrodes, and can be reliably used only for procedures lasting 3–10 min [18]. In order to assess the repeatability and robustness of signals over time, we carried out ECG acquisitions from fabric electrodes during rest conditions, over periods of 30 min per day for a week. Each daily acquisition of 30 min has been split in epochs of 50 s, on each of which, we implemented the power spectral density (PSD). In this way, we obtained 36 PSD curves per day. The Welch method has been applied on ECG values corresponding to each epoch, choosing 512 samples and an overlap rate of 50%.

2) Electromyogram (EMG): Surface electromyography was recorded from the biceps brachii during isometric dynamic contractions. Before placing the electrodes, the skin was prepared according to standard procedures. After shaving, the skin was cleansed with alcohol and slightly abraded with abrasive paste. The GRASS-TELEFACTOR was set with the bandpass filter between 10 and 500 Hz [19] and notch filter at 50 Hz. The National Instruments acquisition card was set with a sampling rate of 2000 Hz. Though we were aware that an oversampling produced no significant benefits in detecting timing and amplitude measurements of the electromyographic signal [20], we chose, however, twice the Nyquist rate, in order to further digitally filter the acquired data. Surface EMG signals were detected with two electrodes, with 20 mm interelectrode distance, in differential configuration with a ground reference electrode placed at the left wrist. The experimental paradigm consisted of alternating resting and muscular contraction periods.

C. Piezoresistive Sensors

A coating process has been applied on jersey fabric, art. Spighetta/5575, composed of 86% polyester and 14% lycra (Milior SpA, Italy). The coating layer is realized with a solution of rubber containing micro disperse phases of carbon, and it is used to realize efficient strain-gauge sensors.

D. Acquisition of Biomechanical Signals Through Sensors

Strain fabric sensors have been used to detect both respiration and motor activity signals (see Fig. 4). In order to evaluate the performance of fabric sensors in monitoring respiration, a comparison with the signal obtained by a dc-coupled standard polymeric piezoelectric transducer has been done. Similarly, the performance of fabric piezoresistive sensors in measuring motor activity has been evaluated by comparing their signals with those simultaneously acquired by commercial electrogoniometers. Respiration signal was acquired by a National Instruments (PC-6036E) acquisition card with a sampling rate of 1000 Hz. Movement signals derived from piezoresistive sensors were conditioned by a dedicated electronic card based on a voltage divider configuration and gathered by the acquisition card. At the same time, signals from electrogoniometers were acquired through a serial port by a commercial system from Biometrics.

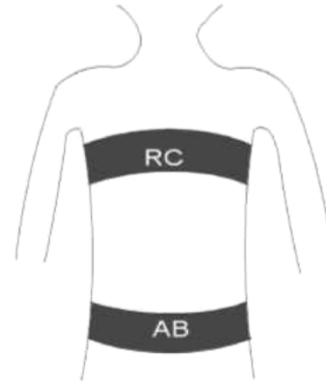

Fig. 5. Piezoresistive sensors located at RC and AB.

1) Respirogram: Respiratory changes (inspiration and expiration) were detected by strain fabric sensors placed along the semicircumference of the rib cage (RC) and the abdomen (AB) below the xiphoid process of the sternum (see Fig. 5). Signals derived from both fabric and polymeric piezoelectric dc-coupled transducers were processed. In particular, waveform morphology (respiratory cycle: inspiration and expiration phases) and respiratory frequency (calculated as the number of respiratory cycles per minute) were studied.

2) Movement Sensors: The fabric sensor's and electrogoniometer's responses, both placed on the right elbow joint, were acquired during the flexoextension mevements of the arm. In particular, a subject volunteered to wear both sensors, and he was asked to repeatedly bend his arm, endeavoring to keep the velocity as constant as possible. The output signals were sinusoidal waves in rough. Two trials were monitored. In the first one, the volunteer was required to slowly bend the arm. The output signals, in this case, resulted in an oscillation at approximately 0.5 Hz. In the second trial, the volunteer was asked to increase the velocity, giving rise to an oscillation, *a posteriori* measured, of approximately 1 Hz.

III. Experimental Results

A. Acquisition of Bioelectrical Signals Through Electrodes

The preliminary characterization of fabric electrodes, for both ECG and EMG signal acquisition, consisted of measuring the electrode impedance in the transversal direction. Electrodes exhibit a pure resistive behavior, with a mean resistance value of 0.03 Ω/cm^2 in the frequency range from dc to 10^6 Hz, which covers the whole working range. This is a reassuring result, because the low contact impedance is crucial for acquiring the most artifact-free electrophysiological data. The impedance value recorded from fabric electrodes is similar to that obtained from Red Dot electrodes in the same frequency range.

1) ECG: In order to verify whether the ECG signal from fabric electrodes deteriorates over time, a daily acquisition

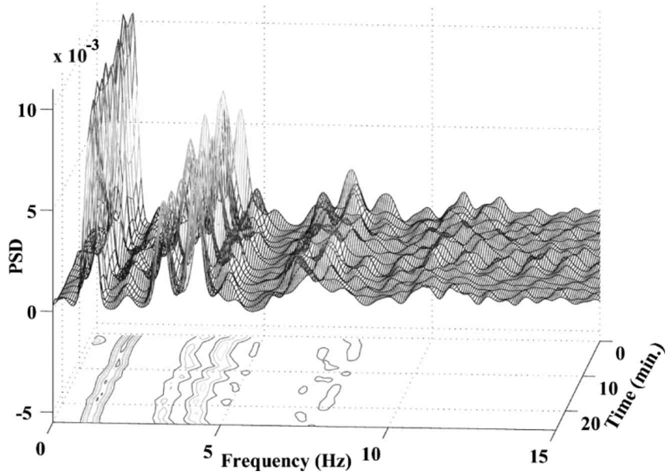

Fig. 6. 3-D PSD curve of ECG evaluated on the first day of the week of acquisition over a period of 30 min, calculated every 50 s. The x axis reports the frequency, the y axis the time, and the z axis the PSD amplitudes. The gray scale is related to the amplitude values. On the xy plane, the isofrequency PSD values are reported.

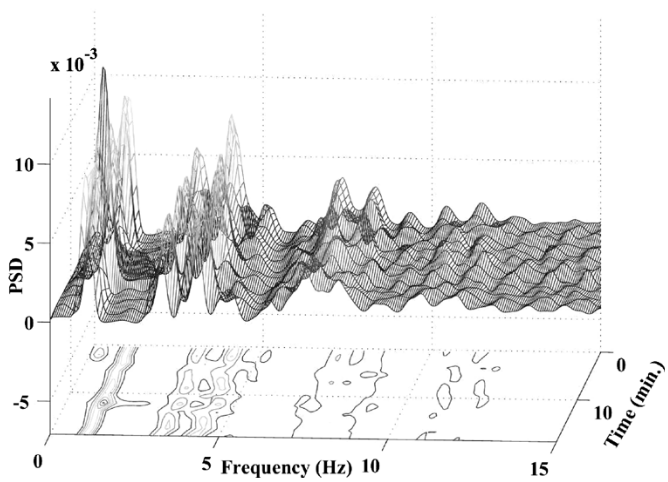

Fig. 7. 3-D PSD curve of ECG evaluated on the last day of the week of acquisition over a period of 30 min, calculated every 50 s. The x axis reports the frequency, the y axis the time, and the z axis the PSD amplitudes. The gray scale is related to the amplitude values. On the xy plane, the isofrequency PSD values are reported.

lasting 30 min was carried out for a week. In addition to a visual inspection of ECG waveforms, which, however, did not show significant differences in the morphology, we also calculated and compared the PSD curves over a period of 30 min, calculated every 50 s, thus obtaining 36 curves. Figs. 6 and 7 show the three-dimensional (3-D) PSD curves of ECG versus frequency and time acquired on the first and last days of the week. It is worth noting that morphology of the PSD curves versus frequency does not significantly change with time, implying that the fabric electrode response did not show any polarization effect. Afterwards, a comparative study in terms of performance, in both time and frequency domains, between fabric and Red Dot electrodes was done. Time analysis has been performed by extracting from an ECG signal a whole cardiac cycle with diastolic and systolic phases. Already, at a glance, the waveform morphology of ECG signals derived from both fabric and standard clinical electrodes, depicted in Fig. 8,

TABLE I
TEMPORAL PARAMETERS OF CARDIAC CYCLE

	Red dot		Fabric	
	Time ms	Amplitude mV	Time ms	Amplitude mV
PR end	126	-0.019	130	-0.032
R	179	0.411	178	0.446
RA		0.43		0.478
J	228	-0.002	230	-0.011
J+60	288	0.019	290	0.021

appears very similar. The visual similarity was quantitatively assessed by comparing the most significant components segments of both signals. Numerical values, reported in Table I, exhibit marginal differences, electrocardiographically irrelevant. In order to further validate similarities between the two electrodes, a careful analysis in the frequency domain has been done. Fig. 9 shows the comparison between the PSDs evaluated for both electrodes. The analysis has been carried out on ECG signals acquired over 5-min sequences by using Welch's method. Even in this case, the two curves are nearly overlapped and the main frequency components are the same. Moreover, the magnitude squared coherence function calculated on ECG signals acquired with fabric and Red Dot electrodes showed a cross-spectrum factor greater than 0.95.

2) EMG: In Fig. 10, the surface EMG signals obtained during the stimulation of biceps brachi, by simultaneously using the fabric and the Ag-AgCl standard electrodes, are shown. Both EMG signals over time coming from fabric and standard electrodes are very similar in terms of amplitude values and time intervals. A more detailed investigation, by comparing the signals in the frequency domain, has been performed. In Fig. 11, the PSD of both signals is reported. The depression at 50 Hz is due to the notch filter used to reject interference from the electric supply. Usually, parameters of the power density spectrum used to provide meaningful information for the EMG frequency spectrum are the mean, peak, and median frequency. Table II reports these values for both Red Dot and fabric electrodes of the PSD of the surface EMG during a contraction phase. Even in this case, numerical differences are irrelevant.

B. Acquisition of Biomechanical Signals Through Sensors

1) Movement Sensors: The signals acquired from the fabric sensor and electrogoniometer placed on the right elbow joint were acquired with time, and reported in Figs. 12 and 13, during slow and fast movements, respectively. Signals coming from electrogoniometers are reported in arbitrary units, while the response of the piezoresistive fabric sensors was measured in Ohms. It is worth pointing out that our aim was to verify whether the fabric sensor was able to measure the motor activity as the number of movements per unit of time. Therefore, for this purpose, only the frequency behavior was relevant. Figs. 14 and 15 show the comparison of frequency components between fabric sensor and electrogoniometer, respectively, for slow and fast movements. In both cases, the principal frequency components are coincident.

2) Respitrace: Respiratory activity produces volume changes of the RC and AB during the inspiratory (volume

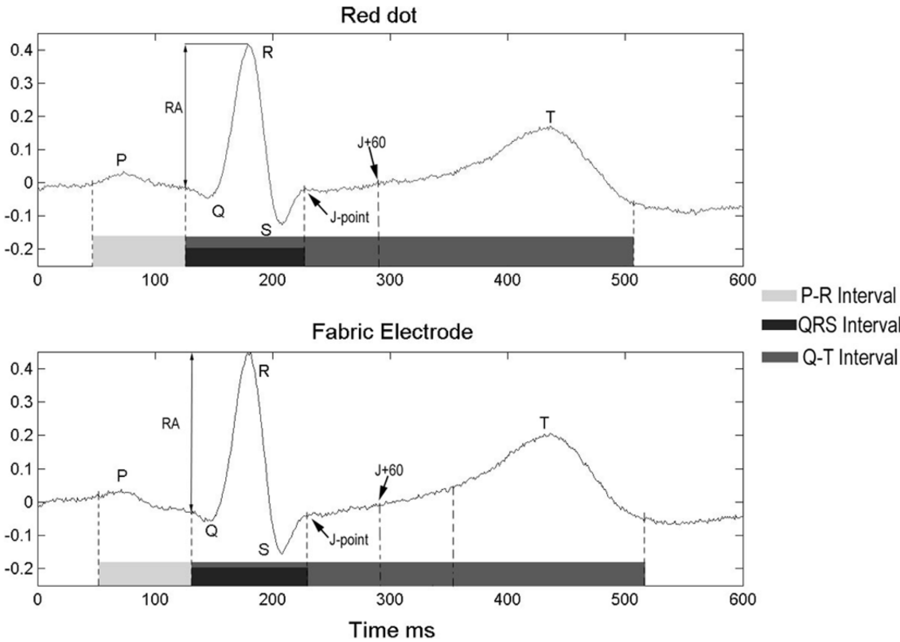

Fig. 8. ECG L1 in rest conditions. P wave, QRS complex and T wave of a cardiac beat is shown for Red Dot (upper) and for fabric electrodes (lower).

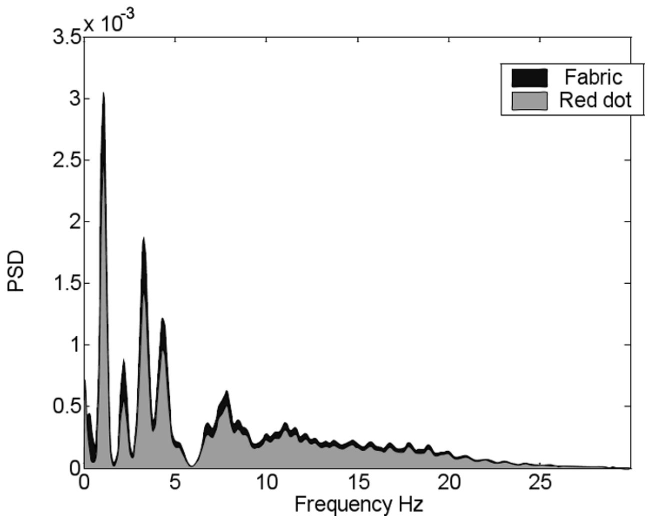

Fig. 9. PSD curve of ECGs. Grey curve refers to ECG coming from Red Dot electrodes, while darker curve is related to ECG coming from fabric electrodes.

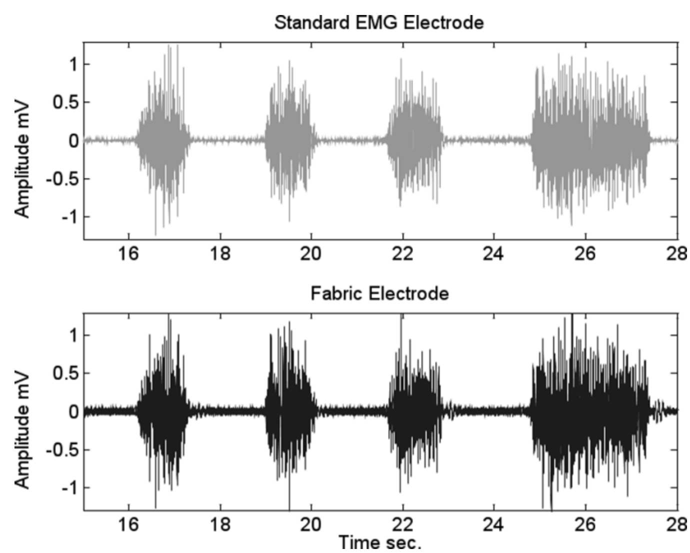

Fig. 10. EMG of the left biceps brachi. Red Dot (upper) and Fabric electrode (lower).

increase) and expiratory (volume decrease) phases of the respiratory cycle. To evaluate the behavior of the fabric sensor, a comparison between the response of piezoresistive fabric and commonly used piezoelectric sensors has been done, and the results are shown in Fig. 16. Fig. 17 shows the respiratory rate changes during a period ranging from slow to fast breathing.

IV. Discussion and Conclusions

In this paper, we reported on a new technology of sensors and electrodes. The innovative idea was to implement transduction functions directly onto textile substrates, which can be integrated in usual garments. Here, fabric sensors able to acquire biomechanical and respiration signals and fabric electrodes to acquire ECG and surface EMG are proposed. In order to assess and validate the performance, signals coming from fabric sensors and electrodes have been compared in time and in the

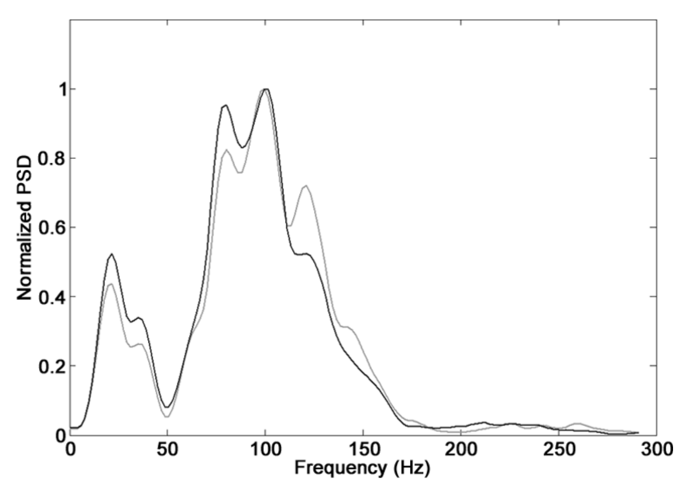

Fig. 11. Normalized PSD of the EMG signal acquired by using Red Dot (lighter dashed curve) and standard (darker continuous curve) electrodes.

TABLE II
FREQUENCY PARAMETERS OF PSD OF THE SURFACE EMG DURING THE CONTRACTION PHASE

	Red dot	Fabric
Mean Power Frequency	99.15 Hz	95.47 Hz
Peak Frequency	99.60 Hz	99.60 Hz
Median Frequency	97.60 Hz	93.75

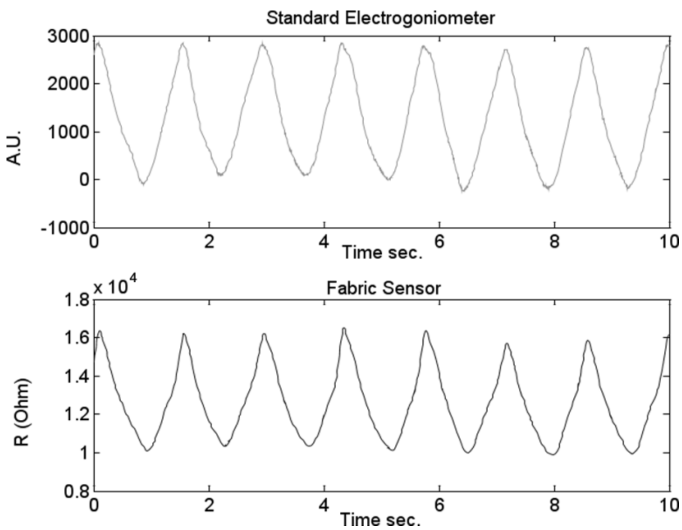

Fig. 12. Comparison between the signals from electrogoniometer (upper) and fabric sensor (lower) during a slow movement of the arm, at approximately 0.5 Hz.

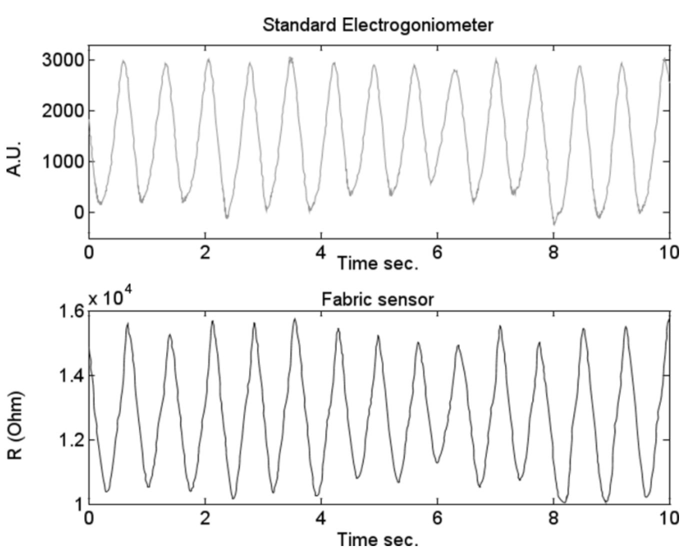

Fig. 13. Comparison between the signals from electrogoniometer (upper) and fabric sensor (lower) during a fast movement of the arm, at approximately 1 Hz.

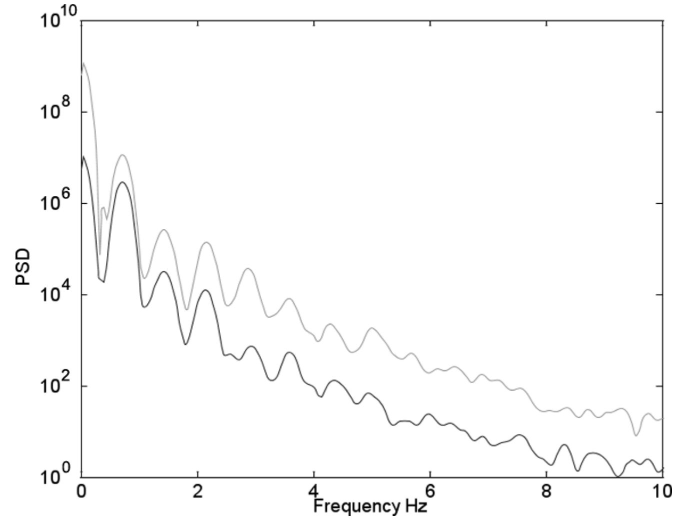

Fig. 14. Comparison between the spectral components of the movement signal acquired simultaneously from electrogoniometer (lighter dashed curve) and fabric sensor (darker continuous curve) during a slow movement of the arm, at approximately 0.5 Hz.

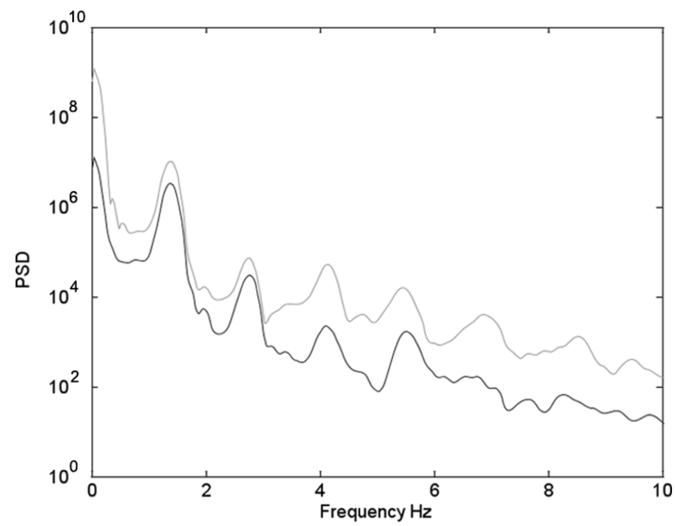

Fig. 15. Comparison between the spectral components of the movement signal acquired simultaneously from electrogoniometer (lighter dashed curve) and fabric sensor (darker continuous curve) during a fast movement of the arm, at approximately 1 Hz.

frequency domain with standard sensors and clinical electrodes. Experimental results showed that fabric electrodes and sensors can be adequately employed to acquire and monitor vital physiological and biomechanical signals, without loss of information. The qualitative similarity of the signals has been assessed by a quantitative analysis in the time and frequency domains of the signals claiming a complete overlap. Although these results are very encouraging, some specific issues should be more carefully addressed when the isolated fabric sensors and electrodes are further developed into an integrated wearable garment. For example, the reproducibility of the positioning of the sensitive spots in the garment represents a critical point. In particular, for ECG, while the positioning of the standard leads (L1-L2-L3) at the root of the arms and at the left groin will not appreciably differ from the standard wrist and foot positioning, the recording from precordial leads has to meet more strict topographical constraints. Another problem is represented by the deterioration of the signal-to-noise ratio produced by micromovements of the electrodes with respect to the skin. So far, this has been neutralized by the interposition of hydrogel membranes, but further efforts in modifying the structure of the yarn in the sensitive spot are necessary to stabilize the skin–electrode contact. Most probably, these problems explain why, so far, in the smart shirts used to monitor vital signals in human subjects, standard clinical electrodes and sensors have been simply inserted in the

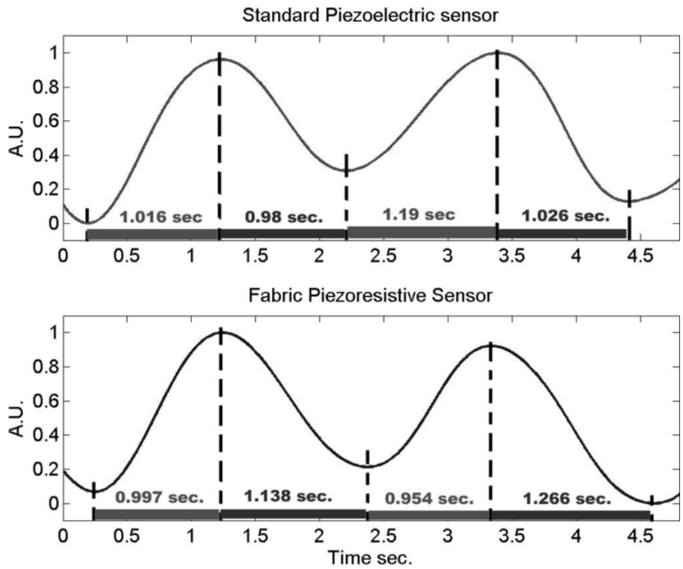

Fig. 16. Respiration signal obtained from a standard piezoelectric sensor (upper) compared with the signal obtained from a fabric piezoresistive sensor (lower).

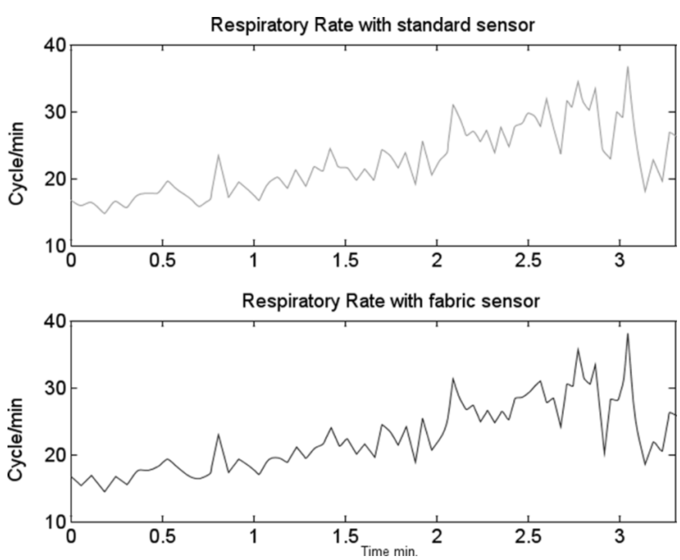

Fig. 17. Respiration rate obtained by computing the distance between the minimal local values for standard piezoelectric sensor (upper) and piezoresistive fabric sensor (lower).

fabric, thus forming a hybrid tool [16]. Our idea was to implement transduction functions directly onto textile substrates employed to construct a comfortable, easily wearable smart garment. This solution appears suitable for several reasons: 1) the garment can be personalized, and this will optimize the positioning of the sensitive regions, thus making the monitoring more accurate and reproducible; 2) such a garment will be more simple and self-manageable; and 3) the absence of visible standard electrodes and connecting wires may contribute to a better compliance of the subjects in accepting this form of monitoring. Finally, any effort in this direction will add a valuable technological contribution to textile engineering, and to its practical applications, such as in telemedicine.

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