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INNOVATION

Electrical characterization of conductive textile materials and its evaluation as electrodes for venous occlusion plethysmography

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Abstract

The ambulatory monitoring of biosignals involves the use of sensors, electrodes, actuators, processing tools and wireless communication modules. When a garment includes these elements with the purpose of recording vital signs and responding to specific situations it is call a 'Smart Wearable System'. Over the last years several authors have suggested that conductive textile material (e-textiles) could perform as electrode for these systems. This work aims at implementing an electrical characterization of e-textiles and an evaluation of their ability to act as textile electrodes for lower extremity venous occlusion plethysmography (LEVOP). The e-textile electrical characterization is carried out using two experimental set-ups (*in vitro* evaluation). Besides, LEVOP records are obtained from healthy volunteers (*in vivo* evaluation). Standard Ag/AgCl electrodes are used for comparison in all tests. Results shown that the proposed e-textiles are suitable for LEVOP recording and a good agreement between evaluations (*in vivo* and *in vitro*) is found.

Keywords

E-textiles, venous occlusion plethysmography

History

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1. Introduction

In the past years a new tendency in patient and athlete monitoring has emerged. The huge advance in miniaturization techniques for electronics devices, the development of electrical textiles and the standardization of communication protocols for WBAN (Wireless Body Area Networks) have made the ambulatory monitoring of vital signs and biopotentials possible [1] and it has become an area of increasing interest for a wide variety of applications in the research field [2–4].

The ambulatory monitoring involves the use of sensors, electrodes, actuators, processing tools and wireless communication modules, with the ability of being integrated into clothes or home articles. When a garment includes these elements with the purpose of recording vital signs and responding to specifics situations, it is often called a 'Smart Wearable System' [5,6].

Sensors and electrodes used in that kind of systems must have the ability of unobtrusively monitoring vital signs. Comfort, biocompatibility and washability are also desirable properties for them. Some types of conductive textile materials (e-textiles) fulfil these requirements and have already been used by several research groups [7,8]. E-textiles and conductive textile materials terms are used as synonyms in this paper.

Much effort is spent in e-textiles characterization and the study of their performance as electrodes and sensors. In this context Kim et al. [9] have developed a textile integrated sensor shirt and evaluated the sensors performance in situations with different values of contact pressure and moisture conditions. Signal-to-noise ratios, from ECG recorded on human skin, were calculated as a quantitative measure. Rattfält et al. [10] have studied the properties of three different conductive yarns and textile electrodes also in contact with human skin. They have investigated how the resistance of a single thread varies with thread length and applied weights. These measurements were performed at five different frequencies. They have also evaluated the electrode polarization potential, its variation and the electrode impedance. Cho et al. [11] have examined textile electrodes properties, comparing them with standard Ag-AgCl electrodes using evaluation of human surface ECG. Other authors have achieved human skin variability independence by means of skin properties emulation using especially designed experimental set-ups. Within this context, Priniotakis et al. [12] have used an electrochemical cell simulating the body/textile electrode system. The effect of textile structure, presence of sweat, sensor size and long-term stability were studied. Beckmann et al. [13] have presented another experimental set-up, designed with a special skin dummy for contact between electrode and skin evaluation.

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This work aims at implementing an electrical characterization of conductive textile materials and an evaluation of their ability to perform as textile electrodes for lower extremity venous occlusion plethysmography (LEVOP) recording.

With this purpose two experimental sets are developed and LEVOP measurements are performed in five healthy volunteers using textile electrodes. The first experimental set is used for determination of conductive textile material sheet resistance variation (SRV) vs fabric length. Four electrode (tetrapolar) impedance spectroscopy measurements are carried out for each e-textile using a frequency response analyser (Solartron SI 1250) and an electrochemical interface (Solartron SI 1278, Solartron Group Ltd, Slough, UK).

Since commercially available conductive textile materials are intended to be used as textile electrodes, skin–textile electrode contact impedance evaluation is important. A skin dummy is used in order to evaluate skin–electrode impedance (Z_c) in a repeatable way. This technique was proposed by Beckmann et al. [13] in order to improve test repeatability. In this work an alternative to measurement electrode configuration is presented. Instead of using the bipolar configuration described by the latter mentioned authors, Z_c determination is carried out by means of three electrode (tri-polar) measurements made with the same frequency response analyser described above. In addition, open circuit tri-polar measurements are used to find out how long polarization potential takes to reach stability. This information is useful to know the waiting time necessary for electrode stabilization once this is placed in contact with human skin.

Finally, e-textiles are intended to be used for continuous lower-limb vascular parameters monitoring, working as electrodes in a garment. Hence, their ability to perform as electrodes is evaluated using impedance plethysmographic waveforms recorded from healthy volunteers. These waveforms are compared with others obtained using standard Ag/AgCl electrodes (3M Health Care, Maplewood, Minnesota).

2. Theoretical background

2.1. Skin–electrode contact impedance

Electrodes used for biopotential recording can be thought of as transducers, since they transform a body ionic current into a current carried out by electrons in the lead wire [14]. When a metal electrode is placed in an electrolyte, oxidation and reduction reactions take place; the distribution of local ionic concentration changes and as result a potential known as 'half-cell potential' is generated. When biopotentials are registered using standard electrodes with wet conductive gel, the gel acts as an electrolyte. Several reasons have been given in the past years to encourage the search for alternatives to gel-based electrodes for biopotential recording. Gel incorporated in Ag/AgCl electrodes surface dehydrates over the course of hours and, because of that, this kind of electrode has limited shelf life and is not reusable. Furthermore, dehydration implies that the gel conductivity is lost, which deteriorates electrode impedance, increasing its value; this is not desirable since it can generate noise and other artifacts. Also, some authors have suggested that gel can cause undesired skin dermatological responses and/or bacterial growth [15]. The good thing about using textile electrodes, besides comfort, is that biopotential recording can be carried out without applying any specific electrolyte [16,17]. This can be achieved since large electrode areas can be implemented without patient disturbance. Besides, the lack of electrolytic gel is usually supplied by moisture on the skin (i.e. sweat).

Skin–electrode contact impedance can be modelled as a layered reactive and capacitive structure [18]. The number of layers varies for each particular interface, taking into consideration electrode type and the presence or absence of electrolyte [19].

2.2. Venous occlusion plethysmography

Venous occlusion plethysmography (VOP) is the most common method used to assess limb blood circulation [20] and it is useful in a wide variety of medical conditions. For example to assess the effect of new vasoactive drugs and hormones in humans [21], to measure calf blood flow during exercise [22], reactive hyperemia [23] or cardiac heart failure (CHF) [24], among others. Venous occlusion plethysmography can be equally well applied to the upper or to the lower limb.

The underlying principle of venous occlusion plethysmography is simple. When the limb venous drainage is briefly interrupted, which is achieved by placing a cuff above the knee, arterial inflow is unaltered and blood can enter the extremity but cannot escape. This leads to a linear growth of the limb blood volume over time that is proportional to arterial blood inflow, until venous pressure reaches the occluding pressure value and a plateau is established. Once this happens, if the pressure is released, venous outflow can be recorded and described by a first order exponential given by Equation (1) [25]. This equation corresponds to an exponential plus a constant, where y_0 represents the initial value at $t = 0$ and *Plateau* represents the constant value. Both the arterial inflow and the venous outflow are useful for different medical applications that require the study of peripheral circulation. In this paper, only venous outflow is analysed.

$$Y(t) = (y_0 - Plateau).\exp\left(-\frac{t}{\tau}\right) + Plateau \qquad (1)$$

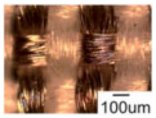
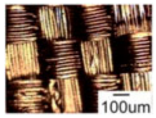
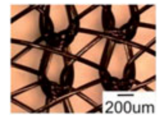
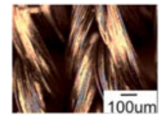
One common mode to record VOP is by means of evaluating volume changes in the limb using impedance plethysmography. This is based on the fact that the body segments impedance reflects the filling state of the blood vessels contained [26]. So, this filling state (blood flow) can be assessed by means of impedance recording, thus a potential difference recording.

3. Materials and methods

3.1. Electrodes

Nowadays a wide variety of textile material with electrical properties similar to those of solid metals is commercially available. Five types of conductive textile materials (LessEMF, Latham, NY) have been chosen for textile electrodes development. Table 1 presents the chosen fabrics and their most important properties. According to the supplier, all fabrics are hand-washables, but washing will eventually degree the coating except on those fabrics that are made of pure metal threads (Argenmesh and Stainless Steel Mesh).

Table 1. Commercially available conductive textile materials and some important features (provided by the supplier (LessEMF, Latham, NY)).

	Argenmesh	Ripstop Silver	Stainless Steel Mesh	Silver Mesh	Stretch
					
Fabric type	–	Ripstop	Knitted	Knitted	Knitted
Coating process	Solid-metal thread	Whole fabric	Solid-Metal thread	Whole fabric	Whole fabric
Base fabric	45% Ny 55% Silver	100% Ny	100% Surgical Stainless Steel	100% Ny	76% Ny 24% Elastic Fibre
Metal coating	100% Ag	100% Ag		100% Ag	100% Ag
Metal purity		>99%		>99%	>99%
Temp. Range	-30 to $90\,^{\circ}\mathrm{C}$				

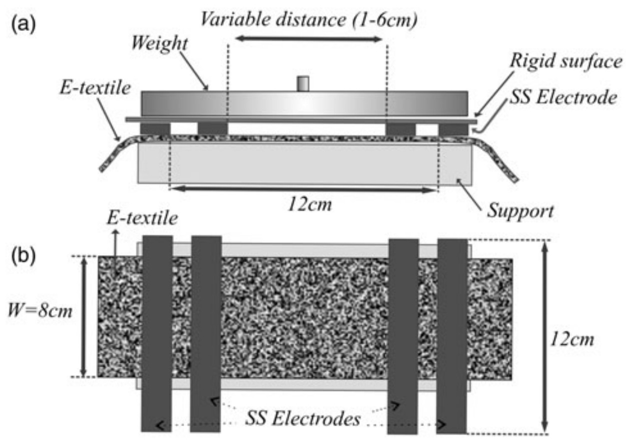


Figure 1. Experimental set-up designed for the evaluation of conductive textile material *sheet resistance* variation versus length. (a) Front view of the experimental set-up where the conductive textile material, a side view of SS electrodes, the rigid surface and the applied weight are schematized from bottom to top. (b) Top view of the experimental set-up. SS electrodes dimensions and their placement over the e-textile can be noticed.

3.2. Experimental set-ups

3.2.1. Conductive textile material sheet resistance

Sheet resistance is a measure of the resistance of thin films that are nominally uniform in thickness and are considered as two-dimensional entities. When the term sheet resistance is used, it is implied that the current flow is along the plane of the sheet, not perpendicular to it. Therefore, this parameter is suitable for electrical characterization of conductive textiles, and in addition it is usually supplied by the manufacturer. However, the study of sheet resistance variation (SRV) in a different fabric section can be useful when a wide number of large-area electrodes, with very similar electrical characteristics, is intended to be elaborated from the same fabric.

In this paper the SRV vs fabric length is studied using a specially designed experimental set. SRV can be a consequence of no contact between conductive threads in different fabric sections or due to fabric deformations (wrinkles) that are not able to be removed, etc. It is important to remark that the variation of fabric electrical properties (i.e. resistivity) measured using the proposed experimental set does not necessarily imply a fabric inhomogeneity as a consequence of the manufacturing process.

Figure 1 shows the experimental set, which consists of an 8 cm width e-textile sample, four 1×12 cm stainless steel (SS) electrodes, a weight and a rigid surface used for weight distribution over the entire system. The e-textile sample rests attached to a plane surface for analysis, while stainless steel electrodes are placed on the fabric for tetrapolar impedance measurements. In this type of measurements, current is driven through a pair of electrodes (outer electrodes) while the potential is measured across another pair (inner electrodes) in order to eliminate or considerably reduce the contact impedance. The inner electrodes are moved from 1 cm to 6 cm apart, while the outer electrodes are fixed at 12 cm apart.

Five types of conductive textile materials are evaluated and five tetrapolar measurements are obtained for each inner electrodes separation, using a frequency response analyser (Solartron SI 1250) and an electrochemical interface (Solartron SI 1278). Finally, sheet resistance (R_{sq}) can be calculated from the measured resistance by using Equation (2).

$$R_{sq} = R_m \cdot \frac{W}{L} \tag{2}$$

where R_m represents the resistance measured for each inner electrodes separation distance (L) and W is the e-textile sample width (8 cm).

3.2.2. Skin dummy–electrode contact impedance and electrode stabilization

The dielectric properties of human skin depend on various factors, which include its physico-chemical characteristics, the composition of intra- and extracellular fluids, psychological and physiological changes occurring in the subject and the condition of health or disease of the skin [27]. Considering this, human skin varies enormously from subject to subject, in different body regions and even at different times. So, in order to improve test repeatability, an agar–agar synthetic skin is used for skin dummy–electrode contact impedance (Z_c) determination. The skin dummy is elaborated following the protocol used by Beckmann et al. [13], where 400 ml of saline solution with a conductivity of $29.3\ \mu\mathrm{S\,cm^{-1}}$ is mixed with 7 g/100 ml of agar–agar. The saline solution conductivity is measured using a hand-held conductivity meter (Oakton Inst., Vernon Hills, Illinois). The resultant skin dummy consists of a $14 \times 14 \times 3.8$ cm gelatinous bulk.

The experimental set used for Z_c determination is presented in Figure 2(a). It consists of an agar–agar skin dummy, a 10×10 cm Stainless Steel (SS) electrode (counter electrode), a textile electrode (working electrode) and a third

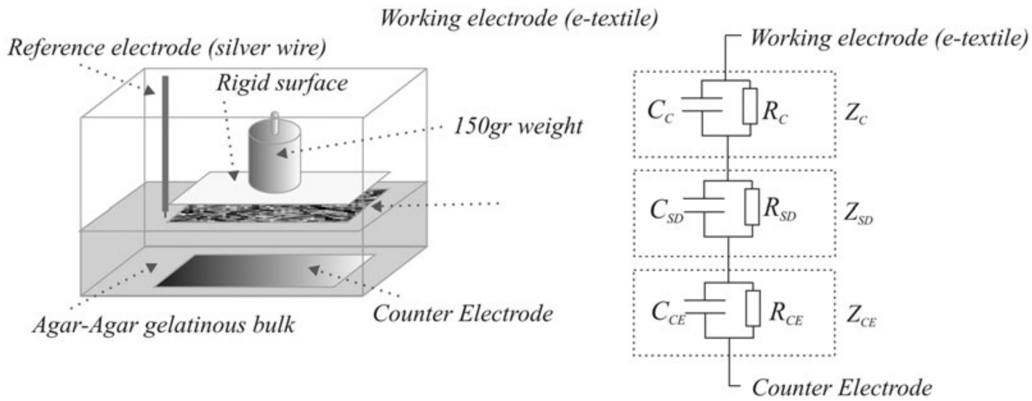


Figure 2. (a) Experimental set-up designed for skin dummy–electrode contact impedance evaluation. The e-textile electrode, the rigid surface and the 150 g weight are located on top of a solid agar–agar gelatinous bulk. (b) Experimental set equivalent circuit.

10 cm silver wire electrode (reference electrode) required for tri-polar measurements. The freshly prepared agar–agar is liquid, so a barrel is needed to contain it. In this work a transparent container (semi-transparent box in Figure 2(a)) has been used, the counter electrode has been placed at the bottom of this and then the agar–agar has been poured into the container. Once the agar–agar solidifies, the working electrode can be placed over it. In the Figure, the agar–agar is the grey part of the semi-transparent box placed between the *counter* and the working electrode. The reference electrode is a silver wire that drills the agar–agar very close to the working electrode.

Figure 2(b) presents the experimental set equivalent circuit. It comprises three RC parallel impedances: Z_C, Z_{SD} and Z_{CE}. Z_C and Z_{CE} represent the impedances that take place at the skin dummy–working electrode and skin dummy–counter electrode interfaces, respectively [18]. Z_{SD} models the skin dummy impedance, where C_{SD} models the bubbles that can result from the agar–agar solidification process and R_{CD} represents the agar–agar alternative resistive path. By means of tri-polar measurements Z_C is measured without the influence of Z_{CE}, which is the advantage of this configuration. When Z_c is measured another impedance that represents the agar–agar section located between the working and the reference electrode appears, but its contribution is negligible.

Several authors have proved that Z_c values vary according to contact pressure applied between the skin and the electrode. Furthermore, lower Z_c values are obtained and test repeatability is improved when contact pressure is increased [13,28]. In this work a 1.5 N force is applied over the textile electrodes in order to get a contact pressure value between the electrodes and the skin dummy that does not change among measurements.

Tri-polar impedance spectroscopy measurements are made with Solartron in order to evaluate skin dummy–electrode contact impedance vs a standard Ag/AgCl reference electrode. Considering that the conductive textile materials could be used as custom-made electrodes in a garment, five 8×8 cm samples from each e-textile were analysed in order to evaluate repeatability. First, the open circuit potential is measured for each textile electrode to evaluate the stabilizing time of the half-cell potential. During this kind of measurements no current flows into the sample. Then, the Z_c is measured, in the range of 1–65 KHz. Each measurement is repeated five times.

3.3. Venous occlusion plethysmography (VOP)

One of the most common methods for measuring VOP involves a constant amplitude alternating current applied to the analysed limb segment and the measurement of the potential difference developed across the tissue [26]. The alternating current frequency range used for measuring VOP goes from 20–100 KHz. In this case, lower extremity venous occlusion plethysmography (LEVOP) waveforms are recorded using a custom-made battery-powered tetrapolar impedance plethysmograph. A 20 kHz alternating current is injected into the lower limb by means of two electrodes, E1 and E2. This current generates a voltage signal whose amplitude is modulated by lower limb impedance changes. This voltage is picked up by another pair of electrodes, E3 and E4, placed between E1 and E2 electrodes. Then the amplitude-modulated signal is recorded and processed (Figure 3a). Impedance changes in the limb, as a consequence of limb blood flow, are expressed as potential difference changes (V [volts]). This is because the intention is to analyse the LEVOP waveform instead of the absolute values of blood circulation.

In order to assess the ability of conductive textile material to perform as electrodes, LEVOP waveforms are recorded from five healthy volunteers with no history of cardiovascular disease using the previously described system. Prior to volunteer participation in the trial, a written informed consent form is signed according to the Declaration of Helsinki.

Three sets of e-textile electrodes—made from Ripstop Silver, Stretch and Silver Mesh materials—and a fourth set of standard Ag/AgCl electrodes, required for comparison, are used. Each set involves the use of four electrodes (E1, E2, E3 and E4). In all cases the size of textile electrodes is 2×21 cm.

The procedure for recording LEVOP includes the following steps: first, the healthy volunteer is positioned in a supine position and the electrodes are placed as shown in Figure 3(a). Before being placed on the skin, the textile electrodes are slightly moistened with water in order to simulate skin perspiration. Second, a cuff occluder is located just above the knee and the volunteer remains at rest for a 12-min waiting period (necessary for electrode stabilization). Third, the voltage value related to the basal impedance is recorded. Fourth, the cuff occluder is rapidly inflated to a 50 mmHg pressure causing limb venous occlusion. This situation is maintained during ~60 s and, finally, the cuff pressure is

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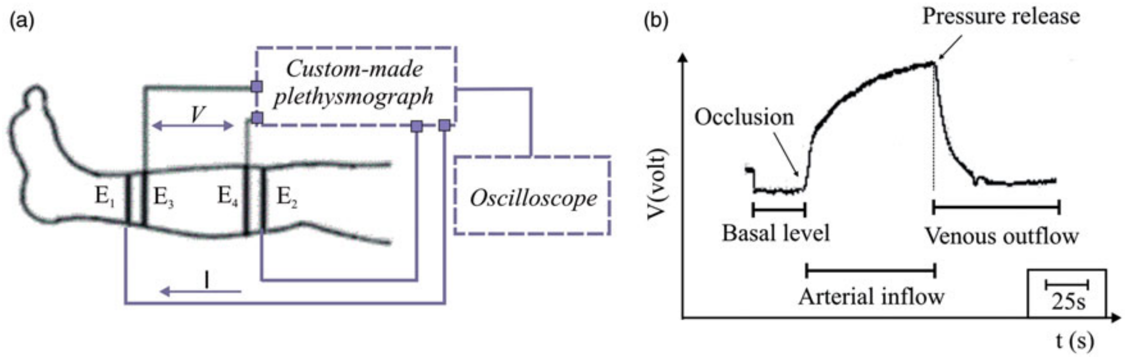

Figure 3. (a) Electrodes arrangement for LEVOP recording. Current injection is carried out by means of the outer electrodes (E1 and E2) while the inner ones (E3 and E4) are used for voltage measurement. (b) LEVOP record as function of time in a 23-year old healthy volunteer.

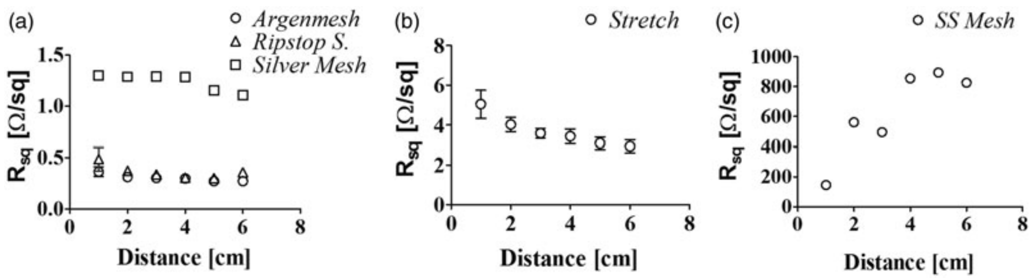

Figure 4. Conductive textile material sheet resistance vs inner electrodes separation distance. These measurements are made using the experimental set-up described in section 3.2.1. Due to the difference in orders of magnitude, results obtained for different textile electrodes are shown in different pictures.

quickly released and the changes in voltage output are recorded (Figure 3b).

Four LEVOP measurements are conducted on each volunteer using each set of electrodes and the custom-made battery powered plethysmograph. The described procedure has been approved by the Ethics Committee of Universidad Nacional de Tucuman (Tucuman, Argentina). All the process is visualized with a digital oscilloscope in volts (V) as a function of time (s) (Tektronix TSD 1001B) (Figure 3b). If limb blood flow increases, limb impedance decreases and so the potential difference does. Hence, the oscilloscope displayed signal is inverted to reflect this blood flow change proportionately.

Figure 3(b) shows the pre-occlusion segment with an approximately constant voltage value, a suddenly increase of the output signal after venous occlusion and, finally, a decay segment that follows the occlusion pressure release. Clearly, occlusion pressure produces a blood accumulation into the limb venous system (arterial inflow); the ratio of the rise of the output signal gradually decreases until the curve reaches a plateau that indicates the maximum venous blood capacity. When the cuff pressure is released, the time-varying output voltage decreases and returns to a value close to the basal one (prior occlusion). The signal segment recorded from pressure release depends on the draining venous capacity and is called venous outflow waveform.

The existence of significant differences between venous outflow waveforms obtained from five healthy volunteers using textile and standard electrodes is studied with a statistics software (GraphPad Prism 5), before this, the waveforms are filtered with a 20-point moving average filter using Matlab v7.9. As Anderson et al. [25] have shown venous outflow curves follow the form of a first-order exponential (1). In this work, a first-order exponential curve fitting is applied to each waveform and then an ANOVA analysis of some relevant parameters of the obtained exponential is performed.

The most significant parameters of the exponential curve (1) are the constant time τ expressed in seconds (s) and the difference between $\underline{y_o}$ and *Plateau*, named *span*, given in volts. If two curves have identical τ, a bigger *span* implies more signal amplitude variation for the same time interval, so this would be more convenient for further analysis.

4. Results

4.1. Conductive textile material sheet resistance

Sheet resistance (Ω/sq)—expressed in ohm/sq—vs inner electrodes separation distance—expressed in cm—is presented in Figure 4. As can be observed, electrodes fabricated from Ripstop Silver and Argenmesh have very similar sheet resistance values and also present the smallest values of all tested e-textile materials. Then Silver Mesh, Stretch and Stainless Steel Mesh electrodes are arranged in increasing order of their sheet resistance. The first two have relatively low sheet resistance values, but these values are slightly higher than those corresponding to Ripstop Silver and Argenmesh electrodes. The only electrode with very large values of the evaluated parameter is the one fabricated from Stainless Steel Mesh.

Sheet resistance variation (SRV), expressed in Ω/sq, is obtained by subtracting minimum from maximun sheet resistance value. Argenmesh, Ripstop Silver, Silver Mesh and Stretch present very small SRV values over the entire measurement range (100 mΩ/sq, 200 mΩ/sq, 200 mΩ/sq and 2.14 Ω/sq, respectively). While Stainless Steel Mesh presents a SRV value of several orders of magnitude greater than that obtained for the other e-textiles (680 Ω/sq). This could be a consequence of the low density of Stainless Steel Mesh and of the rigidity of its threads, which could lead to a poor contact between threads in some fabric sections and to wrinkles formation. All this could affect fabric conductivity and thus modify sheet resistance in some fabric sections.

As Stainless Steel Mesh has the worst performance of all tested conductive textile materials, it is discarded and no longer considered for electrode fabrication.

4.2. Skin dummy–electrode contact impedance (Z_c) and electrode stabilization

The time needed for electrode stabilization has been evaluated by means of open circuit tri-polar measurements. In this work, polarization potential is considered stable when its variation, observed over a period of 60 s, is less than 1 mV. Five samples of each conductive fabric were analysed. The longest stabilization time presented by the e-textiles was 500 s in all cases. Results imply that, in order to avoid errors from polarization potential instability, more than 500 s must be awaited once the electrode is placed in contact with human skin.

Common mode voltage interferences can be reduced and signal-to-noise ratio can be improved by minimizing Z_c values [14,19]. Mean Z_c values over the frequency range (impedance spectrum) obtained for four types of textile electrodes and the ones obtained for the standard Ag/AgCl electrode are presented in Figure 5. Mean Z_c values are calculated from five measurements realized for each sample (five samples) of each e-textile. Only a sample of Ag/AgCl electrodes is evaluated since they are commercially fabricated with very similar characteristics. An impedance spectrum obtained for standard electrode is presented for comparison in both Figures 5(a) and (b).

All tested e-textiles have similar behaviour with regard to the analysed parameter, and Z_c value decreases when frequency increases. This behaviour is not followed by a standard Ag/AgCl electrode, for which impedance values no longer vary from 500 Hz. At low frequencies the interface effects take place and, as result, high Z_c values are observed. Since textile electrodes are dry this effect is significant, which is not observed when wet Ag/AgCl is used. Large differences are observed among textile electrodes regarding absolute values of their impedance spectrums. Looking at Table 1 (section 3.1), and excluding the SS electrode, the metal coating of the other e-textiles is the same (silver coating). Hence, considering their physical properties, the only difference between them is their surface structure.

For the next analysis and considering that Z_c decreases when frequency increases the Z_c values at the lowest and the highest frequency are named 'Maximum' and 'minimum', respectively. 'Range' represents the difference between 'maximum' and 'minimum'. These values can be inferred from Figure 5.

It is remarkable that standard Ag/AgCl electrodes have the higher 'minimum' of all e-textiles. These electrodes show interface effects only at low frequency and then stabilization is achieved. E-textile materials in contact with the skin dummy present stronger interface effects (higher 'maximum' values) than the standard one, which is expected considering that they are dry.

Silver Mesh material presents the highest interface effects and 'range' value. This result is probably related with the low yarn density of Silver Mesh (see Table 1). Ripstop Silver and Stretch textiles have the smallest 'range' values, which could be a consequence of their high fabric yarn density. Nevertheless there is a difference between both; Ripstop Silver is a woven material, while the second one is knitted. Argenmesh textile has a worse performance than Ripstop Silver and Stretch materials, which is expected considering that half of its surface is made of nylon without metal coating.

At this moment, it is important to emphasize that standard Ag/AgCl electrode area is much smaller than the area of analysed textile electrodes. Textile electrodes are far more comfortable than standard Ag/AgCl electrodes, which are fabricated with solid metals. So, when they are intended to be used in a garment, a major area can be implemented.

The results show a high dependence of Z_c with the frequency. Hence, it is important to take into consideration

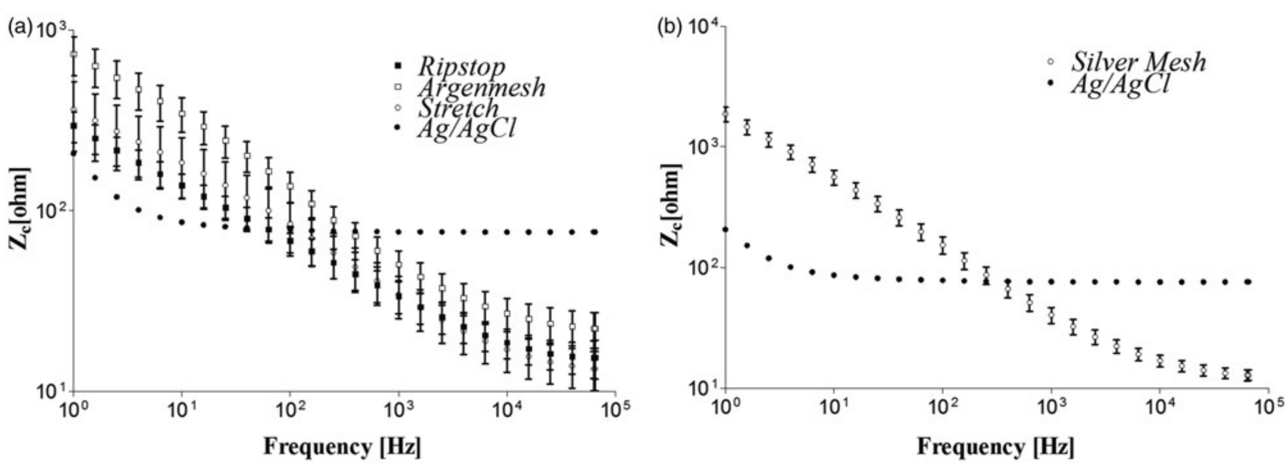

Figure 5. Skin dummy–electrode contact impedance vs frequency evaluated using the experimental set-up described in section 3.2.2. The results are presented in two pictures, (a) and (b), just for better visualization.

that, according to the application—i.e. the working frequency—the electrode could have different performance. As described above, the system used for impedance plethysmography measurements works at a 20 KHz frequency. The Z_c values obtained for 20 KHz frequency for all types of textile electrodes are presented in a box diagram (Figure 6). Silver Mesh and Stretch electrodes have almost identical Z_c median values at 20 KHz frequency (15 Ω) and also the smallest value of all tested electrodes. However, the first one has a smaller deviation. Then, arranged in increasing order according to the evaluated parameter are Ripstop Silver (17 Ω) and Argenmesh (25 Ω). This last electrode has the worst performance of all. The median Z_c value obtained for the Ag/AgCl electrode at a 20 KHz frequency is $76\,\Omega \pm 0.07$, which is higher than the median value obtained for any textile material at 20 KHz frequency. This result is not presented in Figure 6 for scale issues.

4.3. Venous occlusion plethysmography

As mentioned, the aim of this work is to determine if textile electrodes are suitable for LEVOP recording. To do this, the performance of textile electrodes and the standard Ag/AgCl electrode to acquire these waveforms is compared. The waveforms are recorded from five healthy volunteers (three women and two men). A first-order exponential curve fitting (Equation (1)) is applied to each waveform. As an example, the mean venous outflow values, their deviation and their fits obtained from one volunteer are presented vs time in Figure 7(a). All curves are moved so they all start at 1 volt (it is not essentially necessary but it is more comfortable for visual comparison). The goodness-of-fit is evaluated by means of R-square. The R-square obtained, considering all waveform fittings, was 0.980 ± 0.017.

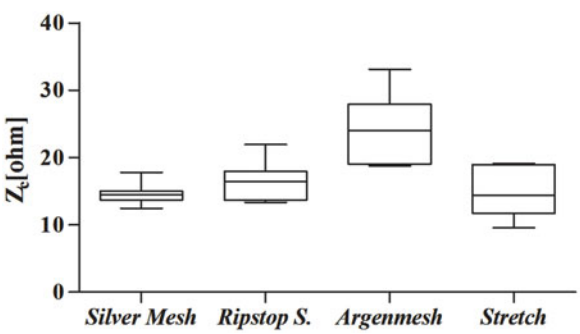

Figure 6. Box plot summarizing the statistics obtained from skin dummy–textile electrode contact impedance evaluation, considering a 20 KHz working frequency.

The parameters τ (s) and *span* (V) obtained from first-order exponential curve fitting are grouped by electrode. A one-way ANOVA analysis is performed for each volunteer; results obtained are summarized in Table 2. This analysis has verified the hypothesis of equality of means (p value >0.05) of τ (s) values among all electrode groups in 80% of the cases. The equality of means implies that no significant differences have been found among electrodes (Silver Mesh, Ag/AgCl, Stretch and Ripstop Silver) regarding to venous outflow recorded within subjects. The situation is different when *span* is considered. In this case the hypothesis of equality of means has not been verified in 80% of the cases. This parameter is related to signal amplitude. So, it can be stated that different amplitude values can be obtained depending on the selected textile electrode.

In Figure 7(b) a box diagram shows the results of *span* statistics for each electrode considering all the volunteers. First, an intra-volunteer span normalization is performed in order to eliminate, in the global analysis, differences related to the subject. It can be seen that venous outflow waveforms obtained using Stretch electrode presents the highest median *span*. Then, in order of decreasing performance are standard Ag/AgCl, Silver Mesh and Ripstop Silver electrode.

Considering only textile electrode, a good correlation is found between Z_c values obtained at 20 kHz (Figure 6)

Table 2. p Values from ANOVA analysis for τ and *span* parameters. These parameters have been obtained from first order exponential curve fitting applied to venous outflow waveforms recorded using textile and Ag/AgCl electrodes.

Volunteer	1st	2nd	3rd	4th	5th
τ analysis	0.0086	0.2403	0.1544	0.8241	0.9895
Span analysis	0.0006	0.0031	0.2099	0.0221	0.0001

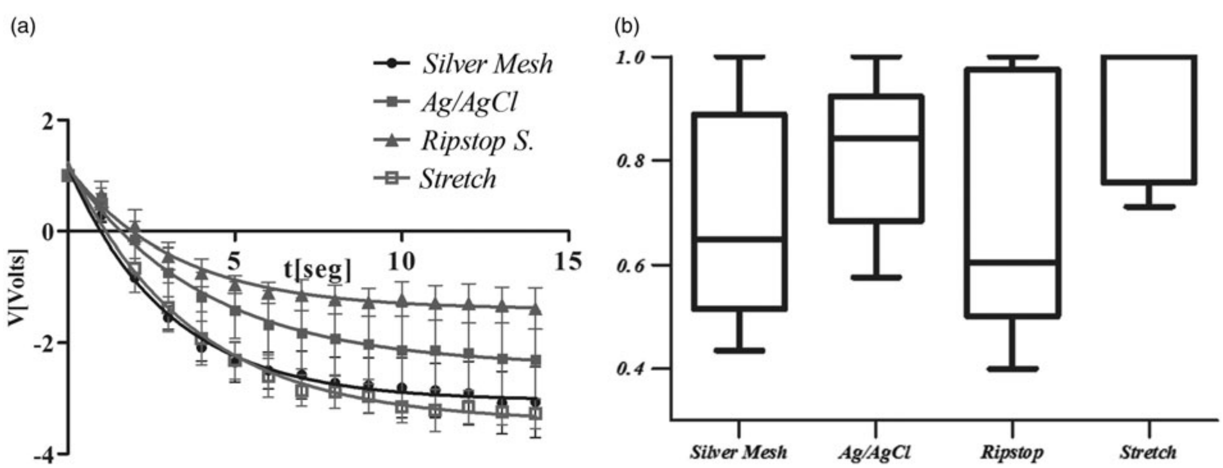

Figure 7. (a) Mean values (in points) and first order exponential curve fit of venous outflow waveforms (in solid line) obtained from one volunteer using three types of textile electrodes and Ag/AgCl electrodes. Four maneouvres are performed per volunteer using each set of electrodes. Whiskers represent the standard deviation. (b) Box plot showing the statistics for 'span' parameter obtained considering all volunteers.

and the median normalized *span* (Figure 7b) obtained for each one. A smaller Z_c value implies a higher *span* value. It is remarkable than Ag/AgCl electrode presents *span* values similar to those obtained for e-textiles while it presents Z_c values much larger than the ones acquired for these conductive fabrics (section 4.2). It could be a consequence of the electrolytic gel applied on the Ag/AgCl surface which improves skin–electrode contact and probably does not improves skin 'dummy'–electrode contact as well.

5. Discussion and conclusion

Several authors have proposed different ways for the characterization and evaluation of conductive textile materials. Usually this characterization is carried out by the analysis of biopotential recorded from the human skin surface (*in vivo* evaluation). However, some research groups have improved test repeatability by means of using experimental set-ups that emulate skin–body properties (*in vitro* evaluation). In this work, a combination of *in vivo* and *in vitro* evaluations is presented. The *in vitro* evaluation involves then use of two experimental set-ups. Conductive textile material sheet resistance variation and skin dummy–electrode contact impedance are evaluated. Results from both evaluations (*in vivo* and *in vitro*) show a good agreement.

Conductive textile material sheet resistance is a parameter that has a direct relationship with the impedance of the entire electrode–body system. An experimental set-up has been designed in order to evaluate the variation of this parameter vs fabric length. When large area electrodes are intended to be fabricated from a fabric, a small variation of fabric electrical properties in a different fabric section is crucial in order to obtain paired electrodes. This parameter has been evaluated by means of measuring sheet resistance variation (SRV) over the entire measurement range. Almost negligible SRV values have been obtained for e-textiles except for Stainless Steel Mesh.

As can be observed in Figures 4 and 6 a low sheet resistance does not necessarily ensure low skin dummy-textile electrode contact impedance (Z_c), as is the case for an Argenmesh electrode. This is due to the difference in textile material surfaces. This material property improves the skin dummy–electrode contact impedance in some cases and in other cases makes it worse. So the Z_c evaluation is also of crucial importance and a second experimental set-up is used in order to do this. Z_c is measured using a tri-polar electrode configuration that includes a third electrode that works as a reference (reference electrode); this electrode is located very close to the working electrode (e-textile). The tri-polar configuration has advantages over the two electrodes configuration presented by Beckmann et al. [13], since the tri-polar electrode configuration measures only the potential changes that occur near to the working electrode with independency of changes that may occur at the counter electrode. Besides, there is no current flowing through the reference electrode, which makes the reference potential very stable.

Finally, to evaluate the e-textile material ability to perform as an electrode, LEVOP records have been obtained from five healthy volunteers. A good correlation has been found between textile electrode performance, regarding venous outflow waveforms records and the Z_c values obtained using the proposed experimental set-up.

Hence it has been shown that the evaluation of the proposed parameters (SRV and Z_c) using the described experimental set-ups is a good way to predict the ability of a textile material to perform as a LEVOP electrode. Also it has been demonstrated that electrodes made with especially selected conductive textile materials—Silver Mesh, Ripstop Silver and Stretch—can be used for LEVOP recording. However, a higher amplitude signal would be obtained by using a Stretch electrode.

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Declaration of interest

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