Dynamic surface electromyography using stretchable screen-printed textile electrodes

Dynamic Surface Electromyography Using Stretchable Screen-Printed Textile Electrodes

Andrea Spanu, *Member, IEEE*, Alberto Botter, *Associate Member, IEEE*, Andrea Zedda, Giacinto Luigi Cerone, *Member, IEEE*, Annalisa Bonfiglio, and Danilo Pani

***Abstract*— *Objective.* Wearable devices have created new opportunities in healthcare and sport sciences by unobtrusively monitoring physiological signals. Textile polymer-based electrodes proved to be effective in detecting electrophysiological potentials but suffer mechanical fragility and low stretch resistance. The goal of this research is to develop and validate in dynamic conditions cost-effective and easily manufacturable electrodes characterized by adequate robustness and signal quality. *Methods.* We here propose an optimized screen printing technique for the fabrication of PEDOT:PSS-based textile electrodes directly into finished stretchable garments for surface electromyography (sEMG) applications. A sensorised stretchable leg sleeve was developed, targeting five muscles of interest in rehabilitation and sport science. An experimental validation was performed to assess the accuracy of signal detection during dynamic exercises, including sit-to-stand, leg extension, calf raise, walking, and cycling. *Results.* The electrodes can resist up to 500 stretch cycles. Tests on five subjects revealed excellent contact impedance, and cross-correlation between sEMG envelopes simultaneously detected from the leg muscles by the textile and Ag/AgCl electrodes was generally greater than 0.9, which proves that it is possible to obtain good quality signals with performance comparable with disposable electrodes. *Conclusions.* An effective technique to embed polymer-based electrodes in stretchable smart garments was presented, revealing good performance for dynamic sEMG detections. *Significance.* The achieved results pave the way to the integration of unobtrusive electrodes, obtained by screen printing of conductive polymers, into technical fabrics for rehabilitation and sport monitoring, and in general where the detection of sEMG in dynamic conditions is necessary.**

***Index Terms*—Dynamic surface EMG, PEDOT:PSS, screen-printing, stretchable textile electrodes.**

Manuscript received March 12, 2021; revised July 12, 2021; accepted August 11, 2021. Date of publication August 16, 2021; date of current version August 27, 2021. This work was supported in part by the DoMoMEA Project through Sardegna Ricerche under Grant Programma Operativo Regionale (POR) del Fondo Europeo di Sviluppo Regionale (FESR) 2014/2020, in part by the Priority Axis I "Scientific Research, Technological Development and Innovation," and in part by European Union's Horizon 2020 Research and Innovation Programme–Search and Rescue Project under Grant 882897. The work of Andrea Spanu was supported by the Programma Operativo Nazionale (PON) project "TEX-STYLE" under Grant ARS01_00996 and Grant PNR 2015-2020. *(Andrea Spanu and Alberto Botter contributed equally to this work.) (Corresponding author: Andrea Spanu.)*

This work involved human subjects or animals in its research. The authors confirm that all human/animal subject research procedures and protocols are exempt from review board approval.

Andrea Spanu, Annalisa Bonfiglio, and Danilo Pani are with the Department of Electrical and Electronic Engineering, University of Cagliari, 09123 Cagliari, Italy (e-mail: andrea.spanu@unica.it).

Alberto Botter and Giacinto Luigi Cerone are with LISiN, Department of Electronics and Telecommunications, Politecnico di Torino, 10129 Torino, Italy.

Andrea Zedda is with the Department of Informatics, Bioengineering, Robotics and Systems Engineering, University of Genoa, 16145 Genoa, Italy, and also with the Department of Electrical and Electronic Engineering, University of Cagliari, 09123 Cagliari, Italy.

Digital Object Identifier 10.1109/TNSRE.2021.3104972

I. Introduction

THE integration of unperceivable sensors and biopotential electrodes into fabrics for the development of easy-to-use, smart garments represents an important goal of wearable electronics. In particular, unobtrusive sensing using soft and conformable textile electrodes is the basis for the development of wearable devices to be applied in healthcare (e.g. for diagnosis [1], rehabilitation [2], prosthetics [3] , [4] and muscle activity monitoring [5]) and sport sciences [6], particularly for professional athletes. From this perspective, the most relevant targeted biopotentials are electrocardiogram (ECG) [7] and surface electromyogram (sEMG) [8]. Approaches ranging from conductive yarns (which can be sewed, embroidered, weaved or knitted onto fabrics [9]–[12]) to natural fibers functionalised with conductive inks [13], have been successfully explored. Recently, poly-3,4-ethylenedioxythiophene doped with poly(styrene sulfonate) (PEDOT:PSS) has gained significant attention in the field of textile electronics. Textile electrodes based on the conductive polymer PEDOT:PSS have been extensively studied and employed as an alternative to commercial disposable pre-gelled electrodes, particularly for sEMG [14], [15] and ECG [16] monitoring. With respect to the material properties and fabrication advantages, PEDOT:PSS-based textile electrodes are more effective and convenient than conventional gelled Ag/AgCl electrodes, especially in terms of comfort and unobtrusiveness, because they can be easily and seamlessly integrated into standard fabrics such as cotton [17], [18] and silk [19].

Despite their possible integration into finished garments, polymer-based screen-printed electrodes have intrinsic fragility to mechanical stretch [20], preventing their widespread use for biopotential monitoring. Therefore, previous studies have only used non-stretchable fabrics as substrates for the production of this type of electrode, with excellent results [15], [21], [22]. However, this leads to imposing workarounds to prevent excessive stretch and consequent damage. Consequently, the application of this technology for the development of smart garments can be hampered by further complexity.

In this work, we address this limitation by proposing a method to embed PEDOT:PSS-based electrodes for sEMG into stretchable garments using a simple screen-printing technique. With a unique printing setup, five pairs of electrodes were fabricated on the interior side of a stretchable polyester commercial leg sleeve. The electrodes were positioned to detect the electrical activity of the vastus medialis (VM), vastus lateralis (VL), gastrocnemius medialis (GM) and tibialis anterior (TA). Monitoring of these muscles is required in several applications, ranging from rehabilitation to sport science. An additional pair of electrodes positioned on the rectus femoralis (RF) was fabricated for skin–electrode impedance measurements. The electrodes were validated in ecologic conditions during physical exercise, which is different than other works conducted under static or relatively static conditions [5], [23], [24]. This evaluation is the most critical test bench for electrodes that adhere to the skin because of the conformation to the body provided by the stretchable fabric, i.e. without the aid of adhesive conductive materials. The experimental validation proposed in this study was specifically designed to demonstrate the effectiveness of this approach in conditions suitable for sport and rehabilitation and to identify the limits of the proposed technology.

II. Materials and Methods

A. Leg Sleeve Fabrication

PEDOT:PSS is almost an ideal material for wearable biopotentials monitoring applications because of its relatively high conductivity (both ionic and electronic [25]) and its use in textile-compliant fabrication processes such as ink-jet printing [26], spray coating [27], electrospinning [28], dip coating [29], and particularly, sponge stencil and screen printing [30], [31].

However, when used in conjunction with screen-printing processes, the material is mechanically limited under stretching [20]. To overcome this limitation, the typical fabrication procedure to print on non-stretchable substrates [15] was modified. The chosen polymer-based ink formulation consisted of a blend of ethylene glycol, PEDOT:PSS CleviosTM PH 1000 (Heraeus Holding GmbH, Hanau, Germany) and 3-GlycidylOxyPropyltrimethoxySilane (GOPS) [32] with 25%, 74% and 1% weight, respectively. The solution was mixed using a magnetic stirrer and then sonicated for 20 min to improve its homogeneity and remove undesired clusters that can have developed during the process. The ink was then cured for approximately 50 min at 70 °C to obtain an adequate density for screen printing.

The target commercial leg sleeves for the production of the smart garment were made of 100% polyester, a stretchable and highly breathable fabric commonly used for the fabrication of fitness clothes. To obtain optimal penetration of the ink into the fabric and guarantee the availability of sufficient material on the stretched fibers, we applied 40% of the radial pre-stretch (with respect to a circular electrode with a diameter of 24 mm) to the printing area on the leg sleeve. Pre-stretch mimics the stretching condition when the garment is worn, potentially guaranteeing the retainment of electrode conductivity during use. After screen printing, the garment was dried at 70 °C for 30 min. This process can be repeated multiple times, adding more layers of conductive ink, to achieve enhanced conductivity and resistance to stretch. In this work, we tested up to three layers of ink for the production of our samples. For screen printing, a polyester screen with 43T mesh delimited by a wax stencil was used, and a rubber squeegee was used to transfer the ink through the mesh to the fabric.

The connection to the recording instrument was obtained using a two-thread stainless steel conductive yarn that was sewn to the final part of the textile electrode and terminated from the opposite side with a medical-grade stainless steel snap button (Spes Medica Srl, Genoa, Italy).

B. Validation Protocol

Five healthy male volunteers (34 to 40 years old) participated in this study. They had no history of neurological or musculoskeletal impairment and diseases. The study was conducted in accordance with the Declaration of Helsinki and the informed consent was obtained from all participants after receiving detailed explanation of the study procedures. Measurement sessions were performed in Cagliari (Italy) in a gym with professional equipment, under the supervision of expert trainers.

The validation of textile electrodes was based on the comparison between EMG signals simultaneously detected, from the same muscle, by off-the-shelf Ag/AgCl and textile electrodes. This approach is based on the hypothesis that the two pairs of electrodes detect the same physiological signal, hence any differences between the recorded signals may be attributed to the characteristics of the recording system. Since the acquisitions were simultaneous, the two pairs of electrodes were placed over different muscle regions. The electrode characteristics and positioning were devised to limit the differences between EMG signal envelopes detected by the two types of electrodes [33]. Participants were asked to shave their right leg the day before the measurement session. Before starting the measurement, the textile sleeve was positioned on the right leg, and the location of the textile electrodes (corresponding to TA, GM, VL and VM) was marked on the skin with a felt tip pen through the fabric. The sleeve was then removed, and the marked regions were gently scrubbed with NuPrep skin preparation paste (Weaver and Co., Aurora, CO, USA). For comparative purposes, a pair of disposable solid-hydrogel Ag/AgCl CDES000024 electrodes (Spes Medica Srl, Genoa, Italy) was positioned close to the marked areas on the skin corresponding to the textile electrodes. For all of the muscles, except the GM, the textile and commercial electrode pairs were aligned on the longitudinal axis of the muscle. Because of the GM morphology, the two pairs of electrodes were placed beside each other, with each pair being aligned longitudinally with the muscle fibres. The commercial electrodes had the same geometry of the textile electrodes, i.e. circular shape with 24 mm diameter, spaced 10 mm to one another. The polyester sleeve embedding the textile electrodes was repositioned and slightly adjusted over the limb to avoid overlap between the two electrode types, which can result in

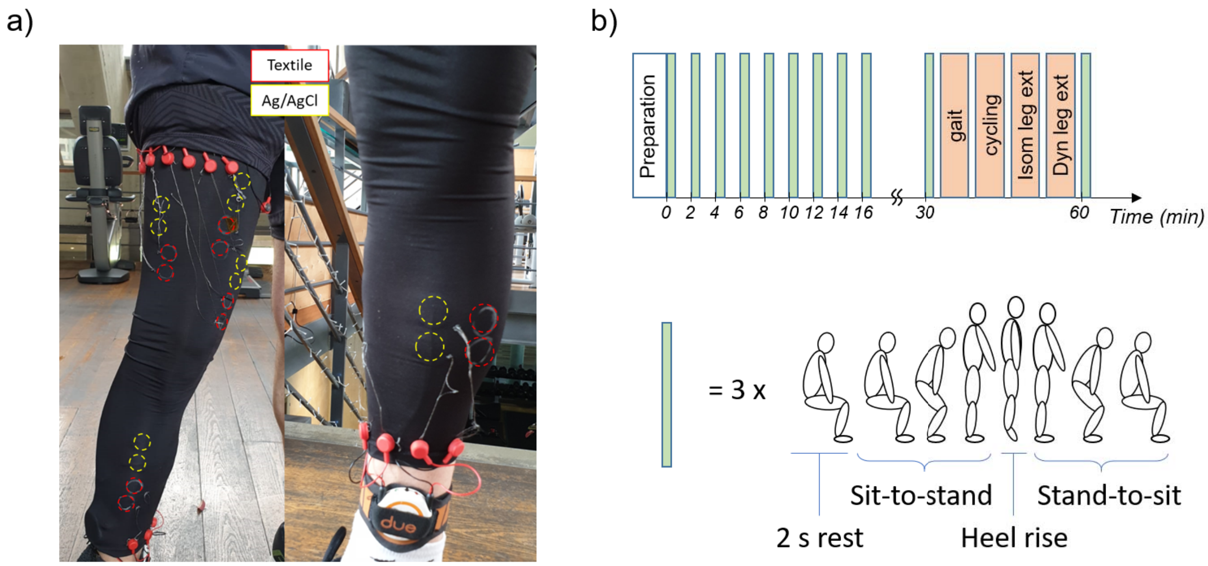

Fig. 1. (a) Front and back of the leg sleeve worn by a participant during the experimental protocol and before the application of the adhesive tape to block perspiration. (b) Textile electrodes are marked in red, whereas disposable commercial electrodes are marked in yellow. Experimental protocol used to test the developed technology.

undesired short circuits. The leg sleeve with the positioning of both the textile and Ag/AgCl electrodes is shown in Figure 1a.

Before starting the measurements, a few droplets (<1 ml) of saline solution were added to the external side of the polyester of the textile electrodes to improve ionic exchange at the electrode–skin interface, increase the active surface of the electrode by filling the recessed parts of its rough surface [7] and reduce the electrical impedance at the electrode–skin interface. The concentration of the saline solution was chosen to emulate sweat (0.1% to 0.4% NaCl [34]). Then, the back of the electrodes was covered with an impermeable and adhesive pad to prevent the region from quickly drying out because of the high breathability of the polyester.

The experimental protocol was designed to quantify the functional performance of textile electrodes in different experimental conditions and evaluate the stability of the electrode–skin contact over time, as schematically depicted in Figure 1b. Participants were asked to perform eleven series of three sit-to-stand movements, including a heel-rise in the stand phase, sitting for two seconds before each sit-to-stand (they started sat down, stood up, performed a calf-rise movement and then sat down again). The series started two min after the saline solution application to enable the stabilisation of the electrode–skin interface, and were performed every two min for the first 16 min ($t_0 = 0$, $t_1 = 2$ min, $t_2 = 4$ min, … $t_8 = 16$ min), and then after 30 and 60 min (t_9 and t_{10}, respectively) from t_0. The electrode–skin impedance between the textile electrodes on RF was measured at t_0, t_5, t_8, t_9 and t_{10}. After the sit-to-stand session, participants performed four physical exercises aimed at dynamically assessing the signal quality during dynamic tasks that are usually performed for lower limbs in rehabilitation, fitness or for neuromuscular assessments. Specifically, participants performed gait (walking at the preferred velocity along a 20 m path), cycling (pedaling on a cycloergometer for 1 min with a cadence of approximately 30 rpm), isometric leg extension (three contractions of four seconds with a knee angle of 120°, with 180° the full extension) and dynamic leg extension (ten extensions from 90° to full extension at approximately 90°/s). The electrode–skin impedance was measured again after the dynamic assessment of the textile electrodes (approximately 60 min from the beginning of the experimental session).

C. Instrumentation for Conductivity, Impedance and sEMG Measures

Prior to the experimental tests on the subjects, the electrodes' material was characterised with bench tests to identify the best fabrication settings for this study. The electrode resistance was measured with a 34411A Digital Multimeter (Keysight Technologies, Santa Rosa, CA, USA). Both the stretched and unstretched electrodes were contacted using a custom printed circuit board with spring contacts (inter-contact distance: 10 mm) pressed onto the fabric with a fixed load (100 g) to ensure reliable and reproducible measurements.

The skin–electrode contact impedance was measured on the RF at 30 Hz with an Agilent 4284 precision LCR meter (Agilent Technologies Inc., Santa Clara, CA, USA). This measurement was performed for each participant only once at the beginning of the experimental session, for the gelled Ag/AgCl electrodes, because of the stability of their skin–electrode contact impedance.

sEMG signals were detected with the DuePro modular sEMG amplifier (OT Bioelettronica, Torino, Italy) and sampled at 2048 Hz (bandwidth 10–500 Hz), with 16-bit resolution. Each module amplifies by 200 and transmits through a Bluetooth link two bipolar sEMG channels. In this study, we used four modules, each of them simultaneously collects the signals from two pairs of textile and Ag/AgCl electrodes applied over the same muscle. DuePro adopts a potential reference system based on the work of Dobrev and colleagues [35], which allows avoiding the use of a physical reference electrode. Two armbands, one fastened in the proximal part of the thigh, close to the upper edge of the sleeve, and another one placed on the ankle, just below the sleeve, were used to hold the four modules (Figure 1a). Signals were displayed and recorded via OTBiolab software (OT Bioelettronica, Torino, Italy). Each couple of both textile and commercial electrodes were connected to a channel of the DuePro using a custom cable with snap button caps, as can be noticed in figure 1A.

D. Data Analysis

sEMG signals were analysed using MATLAB R2016b (The MathWorks Inc., Natick, MA, USA). Before analysis, sEMG signals were digitally band-pass filtered (20–500 Hz, 4th order bidirectional Butterworth filter). For each sit-to-stand series, the time intervals corresponding to the 2-s rest period between consecutive sit-to-stand motions were identified based on the sEMG signals detected by commercial Ag/AgCl electrodes with an amplitude threshold set at 2 standard deviations of the noise. Noise was estimated for each muscle and electrode type by computing the root mean square (RMS) amplitude of sEMG signals detected during the first 2-s rest period. Signal envelopes were obtained by rectifying and low-pass filtering (2nd order bidirectional Butterworth filter, cutoff frequency: 9Hz [36]) the sEMG signals. Although adaptive filtering has been proposed to improve the quality of envelope estimation [37], in this study we used a classic low-pass filter because we wanted to focus on the differences between the two types of electrodes, and the use of a standard technique ensured that those differences depend on the raw signals features and not on filter parameters. The obtained envelopes were then segmented into three portions corresponding to the three sit-to-stand repetitions using the rest intervals previously identified. For each muscle, the normalized Pearson's cross-correlation coefficients between the three sEMG envelopes detected by the textile and Ag/AgCl electrodes were then calculated and averaged to quantify the differences between the recorded signals. The same approach based on the cross-correlation between sEMG envelopes was used to compare sEMG signals acquired with textile and commercial electrodes during gait, cycling, isometric leg extension and dynamic leg extension.

Noise amplitudes were compared using a two-way repeated measures ANOVA (between factor: electrode type). The noise detected by the electrodes at rest over different muscles was considered as independent variables, resulting in 20 observations (4 muscles, 5 participants) at each time point ($t_0 - t_{10}$). A one-way ANOVA with repeated measures was used to study the effect of the time on the cross-correlation between sEMG envelopes simultaneously detected by the two electrode types. The comparison of the sEMG detected by the two electrode types during the four exercises was performed using a two-way ANOVA (variable: cross-correlation between sEMG envelopes). Post-hoc comparisons (Bonferroni's test) were performed when the ANOVA was significant. The threshold for statistical significance was set at $p < 0.05$ for all comparisons.

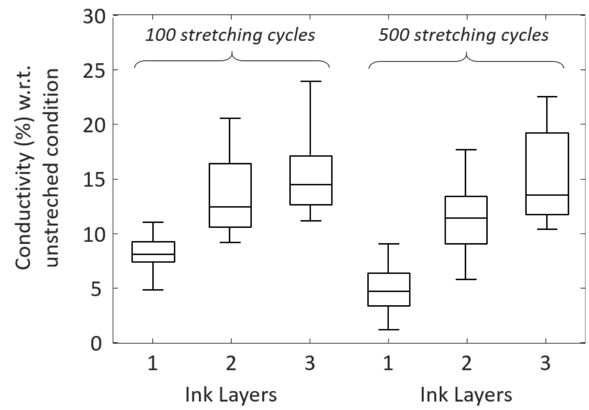

Fig. 2. Surface conductivity (% w.r.t. unstretched condition) of three couples of electrodes developed on a pristine 100% polyester fabric (the same used for the development of the leg sleeve), with 1 layer, 2 layers and 3 layers of conductive PEDOT:PSS-based ink, tested after 100 and 500 cycles.

III. Results

The conductivity of the polyester substrate chosen for the tests, functionalised with the envisioned approach, was assessed on fresh electrodes, after 100 and 500 stretch cycles (Figure 2). The measurements, repeated on pairs of electrodes with one, two or three layers of conductive ink, revealed that three layers yielded the best conductivity. The electrical conductance of the functionalised substrates showed a maximum value of 73 ± 11 mS/cm (3 layers, unstretched) and a minimum value of 1.4 ± 0.4 mS/cm (one layer after 500 stretch cycles).

Although the stretching cycles determined a reduction in conductivity for all the tested electrodes (with 1, 2 and 3 ink layers), electrodes with 3-ink layers performed the best and were therefore chosen for the fabrication of the leg sleeve for the experimental protocol.

Figure 3a reports the comparison between the RMS of the noise detected by the textile and Ag/AgCl electrodes. Statistical analysis revealed a significant effect of the electrode type on the noise values ($p < 0.05$), with textile electrodes being affected by a higher noise level and variability between measures. The time from the electrode application did not significantly influence the noise level detected by either electrode type ($p = 0.13$), and no time–electrode interaction was observed ($p = 0.45$).

The module of the electrode–skin impedance at 30 Hz, which was measured immediately after the electrode placement, largely depended on the subject; it ranged from 3 to 20 kΩ for textile electrodes and from 5 to 50 kΩ for the Ag/AgCl electrodes. The 60-min time course of the textile electrode–skin impedance showed different behaviours in different subjects. After 60 min, the impedance modules ranged between 3.4 and 34 kΩ (from −9% to +115% with respect to the initial value at t_0) (Figure 3b).

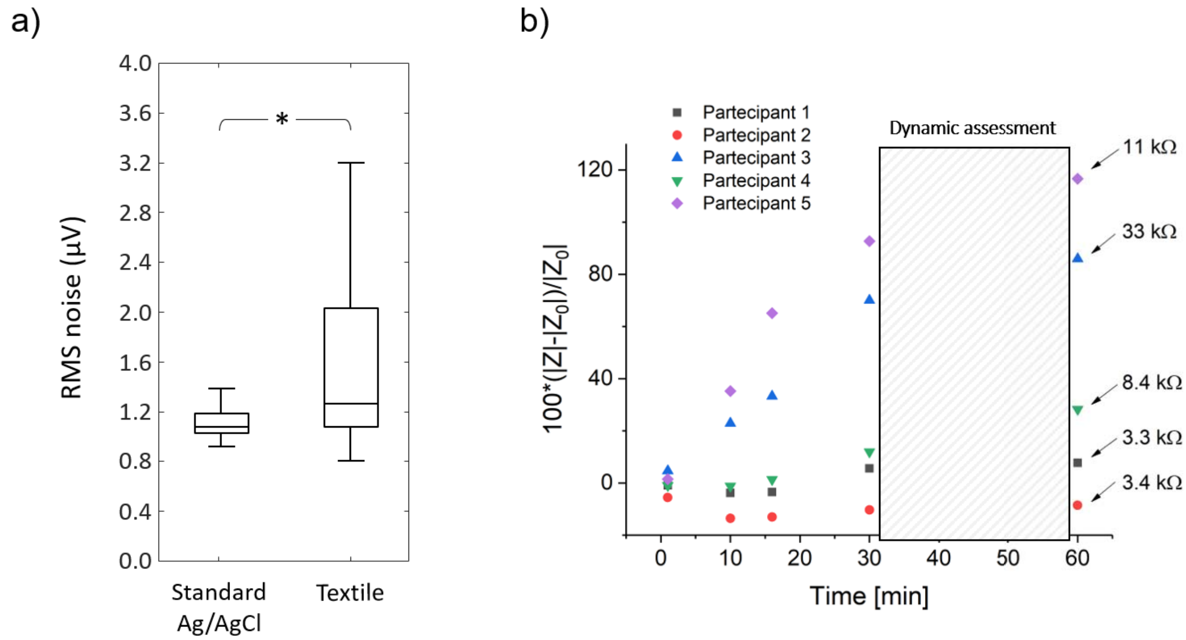

Fig. 3. (a) RMS values of the noise for the different tested electrodes and (b) analysis of the electrode–skin contact impedance, measured between the pair of textile electrodes applied on the RF, during the whole protocol.

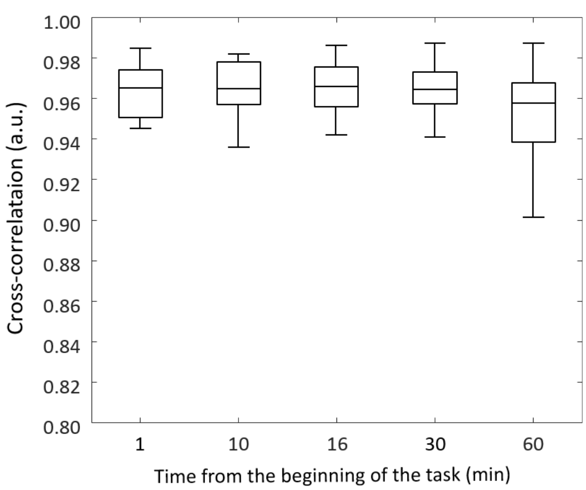

Fig. 4. Cross-correlation between sEMG envelopes simultaneously detected from the leg muscles by the textile and Ag/AgCl electrodes during sit-to-stand and heel-rise tasks.

The cross-correlation between the sEMG envelopes obtained by the two electrode types during the sit-to-stand and heel-rise tasks was always greater than 0.90 (Figure 4). The one-way ANOVA with repeated measures did not show a significant effect of time on the cross-correlation during the time span of one hour ($p = 0.11$), although an increase in the experimental data dispersion is evident for the recording corresponding to 60 min.

Figure 5a shows the cross-correlation values between the sEMG envelopes detected by the textile and Ag/AgCl electrodes for different muscles during the four exercises. Statistical analysis showed a significant effect of the specific exercise and the different muscles on the cross-correlation values, as well as significant muscle–exercise interaction. Post-hoc analysis showed that lower cross-correlation values were obtained during gait in VM. Figure 5b depicts two examples of EMG signals leading to low and high between-envelope normalized Pearson's cross-correlations during gait: 0.78 for VM and 0.98 for TA. The low normalized Pearson's cross-correlation observed for VM is likely due to the occurrence of movement artifacts that strongly affect the envelope shape of the signal detected by textile electrodes.

IV. Discussion

In this work, we presented the development and testing of textile electrodes in dynamic conditions obtained by screen printing of PEDOT:PSS-based conductive ink on 100% polyester, a stretchable fabric commonly used for technical sport garments. Specific considerations of the methods for the creation of the proposed electrodes require more extensive discussion. The size of the electrode was first chosen to be comparable to the available disposable gelled Ag/AgCl electrodes used for sEMG measurements. Regardless of the material, the electrode size influences the electrical performance of the electrode, primarily because of the reduction of the electrode–skin contact impedance with size [7]. From this perspective, our electrodes were medium sized, with an approximate active area of approximately 4.5 cm^2 [38], and the obtained skin-contact impedance parameters are acceptable for the application. Therefore, adopting a larger area is not recommended, because it determines the reduction of the spatial selectivity of the recordings. Conversely, a previous work suggests avoiding PEDOT:PSS-based textile electrodes with smaller sizes, because the low-quality electrode–skin contact can lead to signal distortions [15]. Contrarily, even smaller electrodes could be chosen if the specific recording

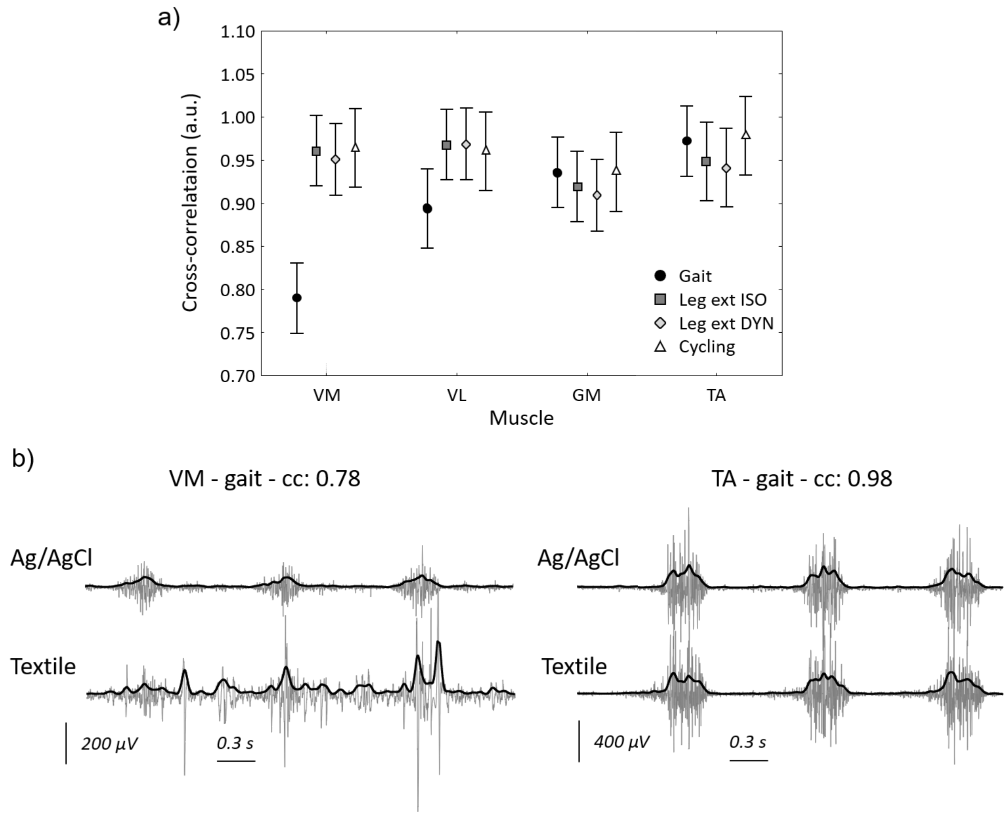

Fig. 5. (a) Cross-correlation between sEMG envelopes simultaneously detected by textile and Ag/AgCl electrodes from VL, VM, GM and TA during gait, isometric leg extension, dynamic leg extension and cycling. (b) Raw sEMG signals detected by Ag/AgCl and textile electrodes during three gait cycles from VM and TA. Envelopes with similar shapes were detected from TA by the two electrode types. The signal recorded by textile electrodes from VM is corrupted by movement artifacts affecting the envelope shape. The cross-correlation value between signals envelopes is reported for both muscles.

task requires it. The versatility of the employed fabrication technology allows for easy adaption of the size of the electrode.

Similar to other polymer-based textile electrodes, our electrodes require a small amount of liquid electrolyte to optimise the ionic transfer between the skin and electrode, which can be obtained by adding saline, using the electrodes in intense exercises or inducing perspiration [13], [39] and keeping it in the electrode proximity with other materials applied on the back of the electrode [40]. Compared with these approaches, the one adopted in this work is closer to Soroudi *et al.* [39], in which a simple polypropylene tape (Müroll GmbH, Frastanz, Austria) was applied to the back of a silver-plated knitted fabric (Shieldex P130, Statex, Germany) to act as a humidity barrier. This solution is simpler and more effective than the approaches based on the application of foam-based reservoirs or rigid anti-perspirating materials, allowing anything to be sewn to the fabric (because polyurethane or polypropylene can be easily laminated on the back of the electrodes for industrial production). Nevertheless, further research is needed to improve this aspect without additional materials on the external part of the garment.

The adoption of multiple layers of conductive ink to improve the electrical properties of a substrate is widely known in the development of organic electronic devices [41], [42]; however, it was not observed in the literature of textile electrodes. Remarkably, the procedure of adding multiple layers of ink improved the robustness of the electrodes to mechanical stress (e.g. stretch) without significant complications in the production process. Resistance to stretch and bending of the textile electrodes is a critical aspect, even though it is seldom discussed in literature. Because of the fabrication process and materials, this problem is often encountered in the electrodes obtained from printing conductive ink onto the fabric surface rather than those created using conductive threads [43].

Despite abundant literature on smart textiles [44], systematic studies on the validation of textile electrodes for sEMG in dynamic conditions are very limited. An interesting attempt was made by Colyer *et al.* [45]: the study participants were asked to perform activities, such as running, cycling, walking and squatting, in a natural environment while wearing sensorised skin-tight shorts to limit relative shifts between the skin and electrodes. Although their technology showed rather good performance, the electrodes (consisting of silver coated yarns) were neither intrinsically stretchable nor easily manufacturable.

The results of our study showed an excellent similarity between the performance of the proposed textile electrodes and disposable gelled Ag/AgCl electrodes, as demonstrated by the analysis of the cross-correlation of the simultaneously acquired signals. Dynamic tests revealed the main influence of the kind of physical exercise and target muscle on the capability of textile electrodes to detect a good quality sEMG signal. Specifically, during gait, large thigh muscles, such as the VM, revealed a reduced level of accuracy of the textile electrodes, probably because of the larger soft-tissue artefacts acting as a low-pass filter on the source signals (Figure 5b). Remarkably, the same electrodes showed good performance before and after gait, suggesting that the problem was due to the specific movement and not a failure of the electrodes themselves. This information could be relevant for the selection of smart garments for the monitoring of those large muscles in gait analysis. Moreover, although an increase of the skin/electrode impedance is clearly observed, the proposed textile electrodes showed very good stability over time along with the ability to reliably monitor the muscle activity during the whole experimental session (1 hour). For two participants (#3 and #5), the increase in the electrode–skin impedance 60 min after the electrode application was quite relevant (80% and 115%, Figure 3b). Although the impedance nearly doubled after 60 min of continuous activity for these two participants, their absolute values did not exceed 33 and 11 kΩ, respectively, which is a reasonably good value for sEMG signal detection.

Despite the limited number of participants, the consistency of the results suggests the usability of our proposed approach to obtain high quality signals during 1-h monitoring of different leg muscles. Nevertheless, because only young male subjects were considered, the generalisation of the results to other subject populations must be done cautiously.

V. Conclusion

In this work, a novel approach to obtain PEDOT:PSS-based screen-printed electrodes on stretchable polyester technical fabrics was presented. The textile electrodes are able to resist normal use and large number of stretch cycles, although further studies are needed to assess also their resistance to washing cycles. The performance evaluation of polymer-based textile electrodes for sEMG in dynamic exercises is fundamental for assessing the actual usability of this technology in real settings, where electrodes must be embedded in stretchable fabrics and must be able to record biopotentials with adequate accuracy. Thus, a rigorous approach to electrode testing was pursued, and the obtained results show that the proposed technology is effective in monitoring sEMG signals from multiple muscles in the lower limb during different motor tasks, with performance comparable with those of conventional disposable gelled Ag/AgCl electrodes. These observations pave the way for the future development of functionalised garments for sport performance assessment or rehabilitation, as the developed technology can reach the objective of a totally unobtrusive and comfortable monitoring of muscle activity.

Acknowledgment

The authors would like to thank the Blue Tribune sport society for the availability of the staff and spaces of the Tribüne Sport Center, Cagliari, Italy, where all the measurements involving human participants took place.

References

[1] X. Xiao, K. Dong, C. Li, G. Wu, H. Zhou, and Y. Gu, "A comfortability and signal quality study of conductive weave electrodes in long-term collection of human electrocardiographs," *Textile Res. J.*, vol. 89, no. 11, pp. 2098–2112, Jun. 2019.

[2] C. Pylatiuk *et al.*, "Comparison of surface EMG monitoring electrodes for long-term use in rehabilitation device control," in *Proc. IEEE Int. Conf. Rehabil. Robot.*, Jun. 2009, pp. 300–304.

[3] S. Lee, M.-O. Kim, T. Kang, J. Park, and Y. Choi, "Knit band sensor for myoelectric control of surface EMG-based prosthetic hand," *IEEE Sensors J.*, vol. 18, no. 20, pp. 8578–8586, Oct. 2018.

[4] D. Tao, H. Zhang, Z. Wu, and G. Li, "Real-time performance of textile electrodes in electromyogram pattern-recognition based prosthesis control," in *Proc. IEEE-EMBS Int. Conf. Biomed. Health Informat.*, Jan. 2012, pp. 487–490.

[5] T. Finni, M. Hu, P. Kettunen, T. Vilavuo, and S. Cheng, "Measurement of EMG activity with textile electrodes embedded into clothing," *Physiol. Meas.*, vol. 28, no. 11, pp. 1405–1419, Nov. 2007.

[6] B. A. Alkner, P. A. Tesch, and H. E. Berg, "Quadriceps EMG/force relationship in knee extension and leg press," *Med. Sci. Sports Exerc.*, vol. 32, no. 2, p. 459, Feb. 2000.

[7] D. Pani, A. Dessì, J. F. Saenz-Cogollo, G. Barabino, B. Fraboni, and A. Bonfiglio, "Fully textile, PEDOT:PSS based electrodes for wearable ECG monitoring systems," *IEEE Trans. Biomed. Eng.*, vol. 63, no. 3, pp. 540–549, Mar. 2016.

[8] L. Guo, L. Sandsjö, M. Ortiz-Catalan, and M. Skrifvars, "Systematic review of textile-based electrodes for long-term and continuous surface electromyography recording," *Textile Res. J.*, vol. 90, no. 2, pp. 227–244, Jan. 2020.

[9] V. Marozas, A. Petrenas, S. Daukantas, and A. Lukosevicius, "A comparison of conductive textile-based and silver/silver chloride gel electrodes in exercise electrocardiogram recordings," *J. Electrocardiol.*, vol. 44, no. 2, pp. 189–194, Mar. 2011.

[10] M. Weder *et al.*, "Embroidered electrode with silver/titanium coating for long-term ECG monitoring," *Sensors*, vol. 15, no. 1, pp. 1750–1759, Jan. 2015.

[11] G. Paul, R. Torah, S. Beeby, and J. Tudor, "Novel active electrodes for ECG monitoring on woven textiles fabricated by screen and stencil printing," *Sens. Actuator A, Phys.*, vol. 221, pp. 60–66, Jan. 2015.

[12] L. Shu, T. Hua, Y. Wang, Q. Li, D. D. Feng, and X. Tao, "In-shoe plantar pressure measurement and analysis system based on fabric pressure sensing array," *IEEE Trans. Inf. Technol. Biomed.*, vol. 14, no. 3, pp. 767–775, May 2010.

[13] R. Castrillón *et al.*, "Electrical performance of PEDOT: PSS-based textile electrodes for wearable ECG monitoring: A comparative study," *Biomed. Eng. Online*, vol. 17, no. 1, p. 38, 2018.

[14] G. M. Paul, F. Cao, R. Torah, K. Yang, S. Beeby, and J. Tudor, "A smart textile based facial EMG and EOG computer interface," *IEEE Sensors J.*, vol. 14, no. 2, pp. 393–400, Feb. 2014.

[15] D. Pani, A. Achilli, A. Spanu, A. Bonfiglio, M. Gazzoni, and A. Botter, "Validation of polymer-based screen-printed textile electrodes for surface EMG detection," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 27, no. 7, pp. 1370–1377, Jul. 2019.

[16] A. Achilli, A. Bonfiglio, and D. Pani, "Design and characterization of screen-printed textile electrodes for ECG monitoring," *IEEE Sensors J.*, vol. 18, no. 10, pp. 4097–4107, May 2018.

[17] A. Ankhili *et al.*, "Comparative study on conductive knitted fabric electrodes for long-term electrocardiography monitoring: Silver-plated and PEDOT: PSS coated fabrics," *Sensors*, vol. 18, no. 11, p. 3890, 2018.

[18] G. B. Tseghai *et al.*, "PEDOT: PSS-based conductive textiles and their applications," *Sensors*, vol. 20, no. 7, p. 1881, 2020.

[19] S. Tsukada, H. Nakashima, and K. Torimitsu, "Conductive polymer combined silk fiber bundle for bioelectrical signal recording," *PLoS ONE*, vol. 7, no. 4, Apr. 2012, Art. no. e33689.

[20] D. Pani, A. Achilli, and A. Bonfiglio, "Survey on textile electrode technologies for electrocardiographic (ECG) monitoring, from metal wires to polymers," *Adv. Mater. Technol.*, vol. 3, no. 10, Oct. 2018, Art. no. 1800008.

[21] H.-S. Cho *et al.*, "Heart monitoring garments using textile electrodes for healthcare applications," *J. Med. Syst.*, vol. 35, no. 2, pp. 189–201, Apr. 2011.

[22] Y. Zhou, X. Ding, J. Zhang, Y. Duan, J. Hu, and X. Yang, "Fabrication of conductive fabric as textile electrode for ECG monitoring," *Fibers Polym.*, vol. 15, no. 11, pp. 2260–2264, Nov. 2014.

[23] G. Paul, R. Torah, S. Beeby, and J. Tudor, "The development of screen printed conductive networks on textiles for biopotential monitoring applications," *Sens. Actuators A, Phys.*, vol. 206, pp. 35–41, Feb. 2014.

[24] O. Tikkanen *et al.*, "Muscle activity and inactivity periods during normal daily life," *PLoS ONE*, vol. 8, no. 1, Jan. 2013, Art. no. e52228.

[25] M. Berggren *et al.*, "How conducting polymer electrodes operate," *Science*, vol. 364, no. 6437, pp. 233-234, 2019.

[26] E. Bihar *et al.*, "Fully printed all-polymer tattoo/textile electronics for electromyography," *Flexible Printed Electron.*, vol. 3, no. 3, Sep. 2018, Art. no. 034004.

[27] F. Gong, C. Meng, J. He, and X. Dong, "Fabrication of highly conductive and multifunctional polyester fabrics by spray-coating with PEDOT:PSS solutions," *Prog. Organic Coatings*, vol. 121, pp. 89–96, Aug. 2018.

[28] Y. Ding *et al.*, "Scalable and facile preparation of highly stretchable electrospun PEDOT:PSS PU fibrous nonwovens toward wearable conductive textile applications," *ACS Appl. Mater. Interfaces*, vol. 9, no. 35, pp. 30014–30023, 2017.

[29] A. Ankhili, X. Tao, C. Cochrane, D. Coulon, and V. Koncar, "Washable and reliable textile electrodes embedded into underwear fabric for electrocardiography (ECG) monitoring," *Materials*, vol. 11, no. 2, p. 256, Feb. 2018.

[30] Y. Guo *et al.*, "PEDOT: PSS 'wires' printed on textile for wearable electronics," *ACS Appl. Mater. Interfaces*, vol. 8, no. 40, pp. 26998–27005, 2016.

[31] S. Sinha *et al.*, "Screen-printed PEDOT:PSS electrodes on commercial finished textiles for electrocardiography," *ACS Appl. Mater. Interfaced*, vol. 9, no. 43, pp. 37524–37528, 2017.

[32] A. Håkansson *et al.*, "Effect of (3-glycidyloxypropyl) trimethoxysilane (GOPS) on the electrical properties of PEDOT:PSS films," *J. Polym. Sci. B, Polym. Phys.*, vol. 55, no. 10, pp. 814–820, May 2017.

[33] T. M. Vieira and A. Botter, "The accurate assessment of muscle excitation requires the detection of multiple surface electromyograms," *Exerc. Sport Sci. Rev.*, vol. 49, no. 1, pp. 23–34, Jan. 2021.

[34] J. Löfhede, F. Seoane, and M. Thordstein, "Textile electrodes for EEG recording—A pilot study," *Sensors*, vol. 12, no. 12, pp. 16907–16919, Dec. 2012.

[35] D. Dobrev, T. Neycheva, and N. Mudrov, "Simple two-electrode biosignal amplifier," *Med. Biol. Eng. Comput.*, vol. 43, no. 6, pp. 725–730, Nov. 2005.

[36] C. Frigo and P. Crenna, "Multichannel SEMG in clinical gait analysis: A review and state-of-the-art," *Clin. Biomech.*, vol. 24, pp. 236–245, Mar. 2009.

[37] S. Ranaldi, C. De Marchis, and S. Conforto, "An automatic, adaptive, information-based algorithm for the extraction of the sEMG envelope," *J. Electromyogr. Kinesiol.*, vol. 42, pp. 1–9, Oct. 2018.

[38] X. An, O. Tangsirinaruenart, and G. K. Stylios, "Investigating the performance of dry textile electrodes for wearable end-uses," *J. Textile Inst.*, vol. 110, no. 1, pp. 151–158, Jan. 2019.

[39] A. Soroudi, N. Hernández, J. Wipenmyr, and V. Nierstrasz, "Surface modification of textile electrodes to improve electrocardiography signals in wearable smart garment," *J. Mater. Sci., Mater. Electron.*, vol. 30, no. 17, pp. 16666–16675, Sep. 2019.

[40] L. Liu, M. Ma, D. Tang, Y. Hu, and H. Liu, "Fabrication and characterization of moisture slow-releasing embroidered electrode and ECG monitoring belt," *Fibers Polym.*, vol. 21, no. 12, pp. 3000–3008, Dec. 2020.

[41] S. Conti, S. Lai, P. Cosseddu, and A. Bonfiglio, "An inkjet-printed, ultralow voltage, flexible organic field effect transistor," *Adv. Mater. Technol.*, vol. 2, no. 2, Feb. 2017, Art. no. 1600212.

[42] H. Shi *et al.*, "Effective approaches to improve the electrical conductivity of PEDOT:PSS: A review," *Adv. Electron. Mater.*, vol. 1, no. 4, 2015, Art. no. 1500017.

[43] K. Arquilla, A. Webb, and A. Anderson, "Textile electrocardiogram (ECG) electrodes for wearable health monitoring," *Sensors*, vol. 20, no. 4, p. 1013, Feb. 2020.

[44] M. Stoppa and A. Chiolerio, "Wearable electronics and smart textiles: A critical review," *Sensors*, vol. 14, no. 7, pp. 11957–11992, 2014.

[45] S. L. Colyer and P. M. McGuigan, "Textile electrodes embedded in clothing: A practical alternative to traditional surface electromyography when assessing muscle excitation during functional movements," *J. Sports Sci. Med.*, vol. 17, no. 1, pp. 101–109, Mar. 2018.