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# Novel Graphene Electrode for Electromyography using Wearables based on Smart Textiles

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**Abstract.** Studies show that biofeedback increases patient proprioception in physical rehabilitation training, improving outcomes; surface Electromyography (sEMG) is particularly appealing as it enables accurate progress evaluation and instant feedback. Furthermore, extending the rehabilitation processes to patients' homes has been shown to increase the quality of the recovery process. This led research to move towards telerehabilitation, however, usability remains an issue in sEMG biofeedback, mainly because of the electrode materials. This work proposes a novel electrode, designed using a Shieldex Technik-TEX P130+B conductive fabric substrate, spray-coated with graphene to reduce the contact impedance with the skin. Experimental evaluation was performed in a population of 16 subjects with ages ranging from 20 to 50 years; results show up to 97% correlation and less than 3dB (in average) degradation of the signal quality comparatively to standard pre-gelled Ag/AgCl electrodes.

**Keywords:** Graphene; Dry Electrodes; Electromyography; Wearables.

## 1 Introduction

In this work we propose and evaluate novel graphene-based e-textile electrodes, to be used in rehabilitation as an alternative to the gold standard. The needs for rehabilitation are increasing; as reported by the World Health Organization (WHO), at least one in every three people in the world will need rehabilitation at some point during their disease or injury. Also, musculoskeletal disorders are typically associated with pain, and limitations in mobility, dexterity, and overall level of function<sup>1</sup>. These conditions may affect joints, bones, and/or multiple body systems [1]. Biofeedback has been shown to improve outcomes in rehabilitation, with surface Electromyography (sEMG) being a particularly appealing modality due to the possibility of objectively measuring muscle functions in real time [2]. Recent work proves that complementing the rehabilitation protocols performed at the clinic, with autonomous work done by patients at home, can accelerate recovery and prevent relapse [3, 4]. The usability of sEMG data acquisition devices is a major barrier, leading to the development of wearables [4].

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<sup>1</sup><https://www.who.int/teams/noncommunicable-diseases/sensory-functions-disability-and-rehabilitation/global-estimates-of-the-need-for-rehabilitation>

Emphasis has been given to smart textiles; however, limitations persist on the signal pickup electrodes, which are single-use and disposable [5].

## 2 Related Work

The popularity of sEMG rehabilitation has led to the creation of highly integrated wearables with dry electrodes. These can constitute one step further in the digitalization and virtualization of human physiology, in particular muscle signals. The Trigno Wireless EMG (Delsys Inc., MA, USA) is an example that uses 99.9% silver (Ag) rods as electrodes, having been extensively used in upper limb assessment (e.g., O. W. Samuel *et al.* in [6]). Still, smart textiles can offer greater convenience and higher usability to end-users [7]. In smart textiles, low electrical resistance is achieved by integrating conductive materials between the fibres of the fabric. This has led researchers to evaluate e-textiles as a way of improving usability of sEMG devices. In [5] Lycra with a surface resistance  $< 1\Omega\cdot\text{cm}^{-1}$  is tested, concluding that a satisfactory performance is achieved in sEMG acquisition comparatively with standard Ag/AgCl electrodes (although with higher baseline noise). The Mbody 3 system (Myontec Oy, Kuopio, Finland) uses silver coated yarns integrated in a garment as electrode; these have already shown promising results in upper and lower body muscle rehabilitation. The topical nature of our research is further reinforced by the recent work of Alizadeh-Meghrizi *et al.* [8], in which conductive silicone rubber yarn is applied to a textile substrate to create a washable smart garment for sEMG.

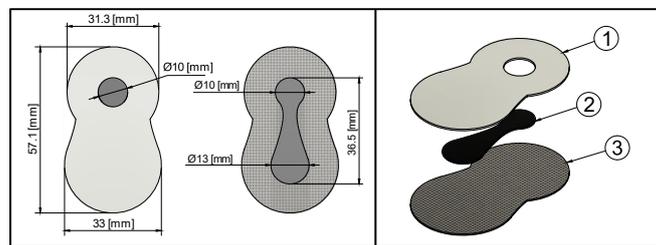
## 3 Proposed Approach

### 3.1 Graphene e-Textile

Graphene has been explored in previous work for producing e-textiles [9-11]; research suggests that electrodes could be made by spraying graphene on a textile, as a way of endowing the material with conductive properties [12]. Typically, non-conductive fabrics are used as substrate, sprayed, or submerged into a graphene solution, conferring conductive properties to the material. Instead, we use Shieldex<sup>®</sup> Technik-Tex P130+B conductive fabric as substrate (Table 1), which is then coated with graphene. This decreases the resistance of the exposed fabric, further reducing the skin contact impedance. Our multilayer e-textile electrode was produced by spray coating, using a 0.3 mm nozzle ejecting air at 3 bars. The graphene-based ink used in this process was supplied and applied by Graphenest<sup>®</sup> (Aveiro, Portugal). As per the details provided by the company, the (wet) graphene content is placed between 5-10 wt.% however, due to intellectual property rights, the exact formulation was not fully disclosed. After the coating, a layer of dielectric material is also applied to expose only a circular surface with a diameter of 10 mm of graphene that will be in contact with the skin (i.e., the electrode). This process was made through screen-printing with a silk mesh fabric of 16t-90/40 Mesh. Fig. 1 depicts the electrode.

**Table 1.** Main electrical, mechanical, and dimensional properties of the standard Pre-Gelled Ag/AgCl electrodes used in regular clinical practice and of the material used in our work.

Property	Pre-Gelled Ag/AgCl	Shieldex® Technik-tex P130+B
Raw material	Conductive self-adhesive hydrogel + Coated polymer	Polyamide (78%) + Elastomer (22%)
Metal plated	Ag/AgCl coated polymer	99.9% pure silver
Metal content		26.5±2% silver
Electrical resistivity	220 $\Omega$	<2 $\Omega \cdot \square^{-1}$
Stretch	None / Polypropylene (PP) transparent substrate	Double stretch / Elongation warp 155-205% / weft 85-125%
Total thickness	1 mm (adapter excluded)	0.55 mm $\pm$ 15%

**Fig. 1.** Proposed electrode dimensional drawing (*left*) and layer forming (*right*). 1. Dielectric screen-printed coating; 2. Graphene sprayed coating; 3. Conductive fabric base substrate layer.

### 3.2 Electrode Assembly

In our envisioned target form factor, the electrode, sensor, and data acquisition system are all placed on a wearable elastic armband. The graphene-based e-textile (Section 3.1) has a thin and flat surface that, depending on the placement with respect to the body surface, may not always ensure a stable contact. Moreover, an interface with the sEMG sensor leads is needed. To address these two aspects, our electrode design includes other mechanical elements beyond the e-textile. With the goal of improving the skin contact, a flexible silicone spacer was designed for placement between the electrode and the elastic armband. In addition, a metallic snap (for textile use) is crimped to the e-textile, enabling to interface the electrode with any standard sEMG sensor lead. Fig. 2 (*right*) illustrates the electrode assembly; as per the SENIAM recommendations for sEMG<sup>2</sup>, two electrodes were sewn to the elastic armband in a fixed position with a 20 mm inter-electrode spacing. The direct connection between electrode and sensor lead in our design, minimizes stretching on the conductive parts of the textile during the armband placement. As seen in Fig. 2 (*centre*), the conductive substrate was partially exposed due to the manufacturing process, thus being a potential noise source. We addressed this issue by applying SYLGARD™ 184 Silicone Elastomer from DOW<sup>3</sup>, which works as encapsulant preventing unintended contact with the skin. This solution was tested by other researchers with positive results [8]. The composite was prepared

<sup>2</sup> <http://www.seniam.org/>

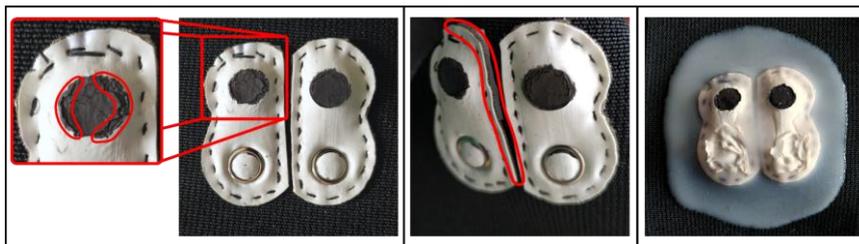
<sup>3</sup> <https://www.dow.com/en-us/pdp/sylgard-184-silicone-elastomer-kit.01064291z.html>

using 10 parts of base compound to 1.2 parts of catalyser, mixed manually for 30 seconds, and degassed at 1 bar of pressure for 120 seconds. Curing was done for 48 hours at room temperature between 20 to 25 °C, the result is shown in Fig. 2 (*right*). It is important to highlight that the electrodes show surface cracks, potentially caused by a poor adhesion between the graphene ink and the substrate (Fig. 2 - *left*). Nevertheless, the work herein presented is a proof of concept, reason for which we decided to continue with the validation of the proposed approach as a way of identifying further improvement points.

## 4 Experimental Evaluation

### 4.1 Materials

An experimental setup was devised for concurrent data acquisition with our electrode and standard disposable electrodes. For the latter we selected the Kendall® H124SG (Cardinal Health, Inc, Dublin, OH, USA), composed of an Ag/AgCl coated polymer; the main characteristics are described in Table 1. These electrodes are primarily designed to monitor ECG signals, hence applicable also to record EMG signals, as demonstrated by Song *et al.* [13]. Two biosignalsplux Electromyography (EMG) sensors were used as sEMG analogue front-end (one for the standard electrodes and another for our graphene-based electrodes). Each sensor has an amplification gain of 1000,  $\pm 1.5$  mV measurement range (with VCC=3V), 25-500 Hz pass-band filtering, input impedance  $>100G\Omega$ , and 100 dB Common Mode Rejection Ratio (CMRR); full specifications can be found in the sensor datasheet<sup>4</sup>. Data acquisition was performed with 16-bit resolution and at 1 kHz sampling rate, using a biosignalsplux Hybrid 8 system (PLUX, S.A., Lisbon, Portugal) through the OpenSignals recording software.



**Fig. 2.** Front view of the electrode (*left*) showing the electrode-skin interface zoomed and rear side of snap connector. Exposed zone (*centre*) of the conductive fabric after application on the band. Electrode after the encapsulant compound deposition (*right*).

<sup>4</sup> <https://support.pluxbiosignals.com/wp-content/uploads/2021/10/biosignalsplux-Electromyography-EMG-Datasheet.pdf>

## 4.2 Protocol

Muscle activation was promoted through an experiment consisting of a single-limb dumbbell arm curl. Participants lifted their forearm against gravity until an elbow flexion of  $90^\circ$  was achieved, and to maintain this position for  $\approx 5$  seconds. This movement was repeated 5 times with dumbbells of 2, 4 & 6 Kg. Similar experiments were already made by Pylatiuk *et al.* [14] while evaluating their wearable device. Electrodes were placed at the *biceps brachii* according to the SENIAM guidelines<sup>3</sup>; although in our study we used the elbow to place the reference electrode instead of the wrist. This was done because the elbow bone is closer to the anatomical location in which our setup is placed, while still providing an electrically neutral tissue [15].

In terms of experimental procedure, first we attach the reference electrode and then secure the armband. Participants maintain a standing position with the armband (with the electrodes) surrounding the *biceps brachii*; the inter-electrode distances are kept at 20 mm following again the SENIAM guidelines. sEMG data is simultaneously acquired using the Kendall<sup>®</sup> H124SG and our graphene-based electrodes, to provide a way to compare both signals in a synchronous way. Each electrode pair is therefore placed at the same distance from the muscle belly to enable a fair comparison.

## 4.3 Population

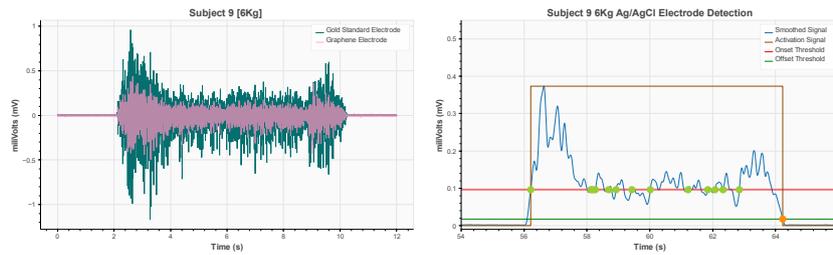
A total of 16 subjects (11 women) were enrolled, ages between 20 and 50 years old, with no associated pathology. Participation was done on a volunteering basis, having as inclusion criteria an age  $>18$  years old. Exclusion criteria included having physical impairments that would be aggravated by the protocol and belonging to a population group considered to be vulnerable. To make the sample more heterogeneous, we have subjects who exercise regularly, alongside people with a more sedentary lifestyle.

## 5 Results

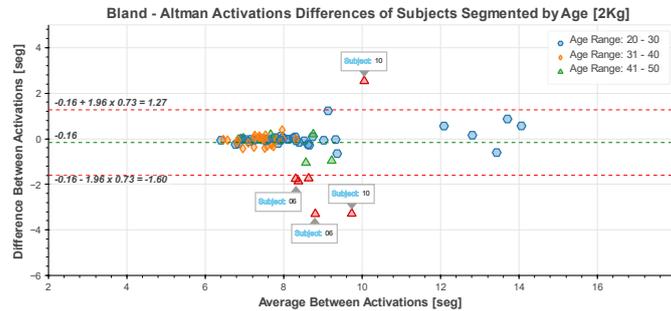
Raw data collected with the experimental setup (Section 4.1) and defined protocol (Section 4.2) was pre-processed to enable an event-based numerical analysis of EMG onset segments. Given that noise can contaminate the signal between the sensor and the acquisition system (e.g., powerline noise introduced through the connection cables), the data is digitally filtered using a 2<sup>nd</sup> order Butterworth band pass filter with cut-off frequencies between of [10; 450] Hz. As a result of both a poor skin-electrode interface of the experimental electrode as well as differences in the input impedance expected by the EMG sensor, which is factory adapted to the disposable electrode impedance to achieve a standardized signal amplitude, we were already expecting that the signal amplitude from our electrode is significant smaller than the gold standard; this effect can be visualized in Fig. 3 (*left*). However, the signal morphology is quite similar. This means that despite the slight variations in the electrode-electrolyte interface, due to differences in the materials in contact with the skin, the signals are still similar.

Onset detection and EMG activation delineation is based on the method by Bonato *et al.* [17]. First the filtered signal is rectified (i.e., the absolute value is determined);

afterwards, smoothing is performed through convolution with a scaled cosine shape Hanning window of 200 milliseconds. The onset and offset detection thresholds are computed as follows:  $Onset_{th} = \mu$  and  $Offset_{th} = \mu - k \cdot \sigma$ . Where  $\mu$  represents the average value of the smoothed signal,  $\sigma$  is the standard deviation and  $k$  is a scale factor to ensure that the  $Offset_{th}$  level is not under the baseline noise level. To bring consistency to the experiment, the scale factor is selected within a range of  $[0.75; 0.9]$ , where the lowest value is applied to the more “baseline noisy” signals and the highest values to the less noisy ones. Delineation of EMG activation segments is performed through binarization of the smoothed signal; 1 is assigned to samples in which  $Onset_{th} < x < Offset_{th}$  and 0 is assigned otherwise. Fig. 3 (right) shows an example of this process. The results of the delineation step were validated through visual inspection, which, despite being a time-consuming task, allows the curation of the EMG activation segments hence providing accurate ground-truth information for comparison. After the curation, the time instants of each EMG activations are determined, and feature extraction is performed. To check if the detections are comparable, we applied a linear regression to the sample recorded for each of the loads used in the study (2, 4 & 6 Kg) between the events detected by both electrodes. Using a Bland-Altman plot, to compare the difference between the activations times is shown in Fig. 4 within the 95% confidence band. We can observe that only 6 of 80 events in total, are out the confidence interval.



**Fig. 3.** Comparison of EMG signal acquisitions (left), where the darker signal (background) corresponds to the standard electrode and the lighter (front) corresponds to our graphene-based electrode. Binarization and detection of the activation (right) showing the acquisition of the gold standard electrode, for the same burst shown at Fig. 3 (left).



**Fig. 4.** Bland-Altman plot of the difference between activation times derived from sEMG acquired with both electrodes, highlighting intra-subject variability examples.

**Table 2.** Average Signal-to-Noise Ratio (SNR) for the standard and proposed electrodes.

Load	Mean SNR for Ag/AgCl $\mu \pm \sigma$ [dB]	Mean SNR for Graphene $\mu \pm \sigma$ [dB]	Mean Scale Factor $\mu \pm \sigma$	EMG Activation Window Correlation (% $\pm \sigma$ )
2 Kg	23.10 $\pm$ 3.48	22.85 $\pm$ 3.61	0.95 $\pm$ 0.09	89.07 $\pm$ 0.05
4 Kg	27.50 $\pm$ 2.23	24.71 $\pm$ 4.43	0.97 $\pm$ 0.05	97.13 $\pm$ 0.02
6 Kg	30.28 $\pm$ 2.16	27.93 $\pm$ 4.66	0.98 $\pm$ 0.05	95.56 $\pm$ 0.03

Through the difference between the number activations inside the confidence interval against the outliers, we can obtain the *detection accuracy* of the prototype comparatively with the gold standard; we obtain a 92.5% accuracy for the 2 Kg experiment, a 97.5% for the 4 Kg and finally 95% for the 6 Kg.

## 6 Conclusions and Future Work

Preliminary results show that the proposed prototype has acceptable performance, nevertheless further work is needed on the manufacturing process, due to descaling of the graphene coating. It is important to highlight that an amplitude attenuation is introduced by the electrode (Fig. 3 - *left*), reason for which the front-end design to acquire the signal needs to be adjusted in future work. On the other hand, besides the similarity in morphology of the signals from both acquisitions, it is mandatory to perform a more detailed study of the frequency content. Future work will be focused on refining and tuning the coating process, seeking to mitigate the surface cracking issues detected in the electrode assembly.

These findings further pave the way for the development of wearable devices based on smart textiles for remote rehabilitation. In this context the medical accuracy is not needed, because the main goal is to give coarse feedback to the patient about the muscle activation during the exercise.

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