


CHARACTERIZATION OF TEXTILE ELECTRODES FOR EMG MEASUREMENTS: IMPEDANCE AND SIGNAL MORPHOLOGY

CARACTERIZAÇÃO DE ELÉCTRODOS TÊXTEIS PARA MEDIÇÕES EMG: IMPEDÂNCIA E MORFOLOGIA DO SINAL

CARACTERIZACIÓN DE ELECTRODOS TEXTILES PARA MEDICIONES EMG: IMPEDANCIA Y MORFOLOGÍA DE SEÑAL


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Abstract

Textile electrodes are an alternative to conventional silver-chloride electrodes in wearable systems. Their easy integration in garments and comfort provided to the user make them an interesting development of textile engineering. The potential of such electrodes to allow more unobtrusive data collection in health and sports context may enable the development of biosensing garments to be used in biomechanics. However, proper validation of the recorded signals is paramount, and few studies have yet presented consistent methodologies for textile-based electromyographic recordings. This study presents the validation of the electrical and morphological properties of electromyographic signals recorded with textile electrode, in comparison to conventional silver-chloride electrodes. Results indicate that both sets of electrodes have identical signal-to-noise ratios, but with distinct impedance frequency responses. Electromyographic envelope morphologies are also identical, although textile electrodes usually have lower amplitudes.

Keywords: Wearable; Textile Electrode; Electromyography; Sensory Garment.

Resumo

Eléctrodos têxteis são uma alternativa aos convencionais eléctrodos de cloreto de prata em sistemas vestíveis. A sua fácil integração em vestuário e o conforto que providenciam tornam-nos um interessante desenvolvimento da engenharia têxtil. O potencial destes eléctrodos para permitir a discreta recolha de dados em contexto de saúde e



desporto poderá permitir o desenvolvimento de vestuário biossensível com aplicação em biomecânica. No entanto, uma adequada validação dos sinais registados é fundamental, e poucos estudos apresentaram metodologias consistentes para registos electromiográficos baseados em eléctrodos têxteis. Este estudo apresenta a validação das propriedades eléctricas e morfológicas de sinais electromiográficos registados com eléctrodos têxteis, em comparação com os convencionais eléctrodos de cloreto de prata. Os resultados indicam que ambos os eléctrodos apresentam razões sinal-ruído idênticos, mas com uma resposta em frequência de impedância distinta. A morfologia do envelope electromiográfico é também idêntica entre electrodos, ainda que geralmente de menor amplitude no eléctrodo têxtil.

Palavras-chave: Sistemas Vestíveis, Eléctrodos Têxteis, Electromiografia, Vestuário Sensorial.

Resumen

Los electrodos textiles son una alternativa a los electrodos de cloruro de plata convencionales en los sistemas portátiles. Su fácil integración en las vestimentas y la comodidad que brinda al usuario las convierten en un interesante desarrollo de la ingeniería textil. El potencial de tales electrodos para permitir una recopilación de datos más discreta en el contexto de la salud y el deporte puede permitir el desarrollo de vestimentas biosensibles para su uso en biomecánica. Sin embargo, la validación adecuada de los señales registradas es primordial y pocos estudios han presentado metodologías consistentes para registros electromiográficos basados en textiles. Este estudio presenta la validación de las propiedades eléctricas y morfológicas de señales electromiográficas registradas con electrodo textil, en comparación con electrodos convencionales de cloruro de plata. Los resultados indican que ambos los electrodos tienen relaciones señal-ruído idénticas, pero con una respuesta de frecuencia de impedancia distinta. La morfología de la envoltura electromiográfica también se identifica entre electrodos, aunque generalmente de menor amplitud en el electrodo textil.

Palabras clave: Sistemas Vestibles, Electrodos Textiles, Electromiografia, Indumentaria Sensorial.

INTRODUCTION

The use of wearable solutions is a popular solution to perform human physiological and movement measurements (TAJ-ELDIN et al., 2018), allowing unobtrusive data collection. This is particularly valuable in biomechanics, where movement should not be impaired by the measuring equipment. Additionally, clinical and sports biomechanics require the availability of measuring solutions that combine convenience with reliability.

Biologically originated electrical potentials such as the electrocardiogram (ECG) and the electromyogram (EMG) are commonly measured on medical and scientific contexts, and they are of great importance to assess cardiac and muscular activity in a myriad of situations. Currently, silver-chloride electrodes are the most frequently used type of electrodes to perform this measurements, allowing the recording of good quality signals (XU; ZHANG; TAO, 2008). However, some disadvantages such as skin irritation because of the fixation adhesives (CATRYSSE et al., 2004), the limited shelf-life, the dehydration over time (GRUETZMANN; HANSEN; MÜLLER, 2007; MAROZAS et al., 2011) and the heterogeneity of coupling silver-chloride electrodes with textile garments, led researchers to start looking for alternatives to record biopotentials.

Textile technology has been providing alternatives to non-conventional sensors (ISLAM; ALI; COLLIE, 2020), especially during long-term and continuous EMG recordings (GUO



et al., 2019). Besides the ability to fully integrate the electrodes into a garment and enable measurements during almost any land-based activity, such electrodes presents other advantages, such as being lightweight (XU; ZHANG; TAO, 2008), and the ability to endure several sessions of use since they can be washed (BIFULCO et al., 2011; COOSEMANS; HERMANS; PUERS, 2005; SCILINGO et al., 2005). Of particular interest for sports biomechanics applications is their flexibility, allowing an improved skin contact, especially in muscles with high curvatures (XU; ZHANG; TAO, 2008).

Despite the possibilities for biomechanical monitoring, textile electrodes are not without some drawbacks. Some challenges have been reported, such as an higher skin-electrode impedance than the silver-chloride electrodes (CATRYSSSE al., 2004), the presence of higher levels of noise (CHO; LEE; CHO, 2009; COOSEMANS; HERMANS; PUERS, 2005) and the higher vulnerability to motion artifacts due to the lack of fixation adhesives. Regardless of the abundance of publications on the development of textile electrodes new designs and fabrication methods, their validation is scarce. Recently, some studies have presented methods for the validation of textile-based EMG electrodes (KIM; LEE; JEONG, 2020; PANI et al., 2019), proving useful methodologies for the reliable assessment of textile products.

This work aims at validating a new set of EMG textile electrodes developed for the integration of functional garments for health and sports applications. To do so, skin-electrode impedance testing, signal-to-noise ratio and the morphological analysis of EMG recordings during isometric tasks with different loads were performed using textile and conventional silver-chloride electrodes simultaneously.

MATERIALS AND METHODS

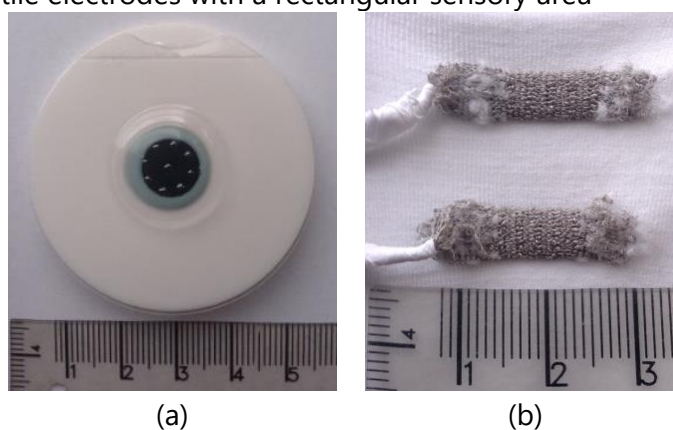
Two groups of healthy volunteers, with no history of neuromuscular conditions or injuries in the upper limbs in the last six months, were recruited from the local university student body. The first group, comprising ten participants (six male and four female) aged 24 ± 3.3 years old, 69.1 ± 12.2 kg of mass, 1.74 ± 0.97 m in height and a Body Mass Index (BMI) of 22.72 ± 1.73 kg/m² participated in the skin-electrode impedance measurements. The second group included ten participants (five male, five female) aged 25.0 ± 3.0 years old, 65 ± 10 kg of mass, 1.69 ± 0.11 m in height and BMI of 22.78 ± 1.76 kg/m², and was involved in the physiological EMG recordings.



The textile electrodes (TE) under validation were a custom-blend of 80% silver plated polyamide (Elitext) and 20% Lycra with an average linear mass of 23.5 tex. Each electrode was manufactured in a rectangular shape with a 1 cm² sensory area. Due to their lack of self-adhesive properties, the TE were sewn on a semi-rigid material (plastazote) and fixed to the skin with an adjustable elastic band. The conventional silver-chloride electrode (CE), used as reference, was the self-adhesive SX-30 model of Dormo&Blayco (Telic S.A., Spain), with a gel film over a 1 cm² circular sensory area. Both sets of electrodes are depicted in Figure 1.

All volunteers were informed of the study objectives and the experimental protocol, following the appropriate ethical procedures in accordance with the Helsinki Declaration and its later amendments. Participants had the opportunity to familiarize themselves with both types of electrodes and the experimental procedures before data collection began.

Figure 1 – The two electrodes used in the study: (a) a conventional silver-chloride electrode, and (b) the EMG textile electrodes with a rectangular sensory area



Source: authors's owns.

Prior to any data collection, the skin in the measuring area was shaven and cleaned with a piece of cotton embedded in a 95% alcohol solution, and slightly abraded until redness with the same cotton piece.

Skin-electrode impedance measurements of both the CE and TE were performed with a Voltage Controlled Current Source based on an Improved Howland Circuit. This circuit generated a 100 μ A sinusoidal current signal from 15 to 1000 Hz, and both impedance magnitude and phase were measured. The magnitude was obtained by measuring the voltage drop between the electrodes and the skin, while the phase shift was calculated as the difference between the current's phase before and after the injection into the body. All measurements



were performed with a NI USB6211 (National Instruments, USA) acquisition board with a resolution of 16 bit and a sampling frequency of 42KSamples/s.

Impedance measurements were performed over the right ventral forearm, with electrodes placed at its midpoint with a 2 cm center to center distance. A thin layer of 0.1 ml of conductive gel (Quick Eco-Gel, Lessa S.A., Spain) was applied on each TE to provide similar skin contact conditions to those of the CE. A settling period of at least two minutes was ensured after placing the electrodes on the skin and a sphygmomanometer cuff was used to maintain a constant pressure of 2.67 kPa (20 mmHg). Impedance was measured from 15 to 1000 Hz, which covered all the frequency range used in the EMG physiological measurements.

The physiological EMG measurements were performed with both sets of electrodes. To this end, a two-stage amplifier was used, amplifying the signal 1100 times (first stage: 100; second stage: 11) with a Common Mode Rejection Ratio (CMRR) over 110 dB. The amplified signals were then recorded using a BIOPAC MP100 (BIOPAC Systems Inc., USA) with a 16 bit resolution and 1000 Hz sampling frequency.

Electrodes were placed on the biceps brachii muscle belly, according to the SENIAM recommendations (HERMENS; FRERIKS; DISSELHORST-KLUG; RAU, 2000), and the reference electrode attached to the elbow. Due to the inability of placing both the CE and TE on the same place, and to prevent substantial differences in the recorded signal, they were carefully placed as near as possible. The CE was placed on the middle of the muscle belly, while the TE was placed distal to CE, along the muscle fibers orientation. Conductive gel was applied in the same manner as in the impedance measurements. The pressure exerted by the TE against the skin was not standardized, but the elastic band was adjusted to keep a good and comfortable skin contact, always by the same researcher.

Participants were asked to sit on a rigid armless chair with the right elbow flexed at 90° and the forearm supine, while keeping the trunk at 90° regarding the chair seat, and the shoulder at 0° of flexion, as described by Oliveira et al. (2009) and Linz, Gourmelon and Langereis (2007). All joint angles were measured with a goniometer. With the participant sitting in the abovementioned position, three maximum voluntary contractions (MVC) attempts with ten seconds of duration and at least two minutes of rest between repetitions were performed with strong verbal encouragement. During each MVC attempt, the maximum isometric force was measured with a Globus Ergo Meter Load Cell (Globus Italia, Italy). The load cell was connected to an ergonomic handle held at the participant's hand, and to the ground, by a



metal chain. The EMG recording obtained during the maximum force production attempt was selected for further signal normalization. After a resting period of at least three minutes, isometric sub-maximal contractions, with the participant sitting with the same posture as during the MVCs, were performed holding dumbbells against gravity, with loads representing 20%, 50% and 80% of the MVC, in a randomized order. For each sub-maximal load, three repetitions of five seconds were performed. A recovery period of at least two minutes between repetitions, and three minutes between loads was observed.

DATA ANALYSIS

Data from the impedance and physiological measurements were processed using Matlab 7.9.0 (MathWorks, MA, USA).

EMG data was digitally filtered with a 20-500 Hz zero-phase Butterworth bandpass filter, followed by a full-wave rectification. The signal envelope was obtained by the calculation of its Root Mean Square (RMS) with a 150 ms window. The resulting envelopes of each electrode were then normalized by the corresponding MVC value. The level of noise content in the signal was measured as the average Signal to Noise Ratio (SNR) obtained from the MVC envelopes of each participant.

The EMG morphological analysis compared the TE and CE envelope patterns while holding different loads. To that end, the Root Mean Square Error (RMSE) between envelopes was calculated as a measure of their overall similarity, the envelope's peak (pEMG) as a measure of their maximum measurement ability, and the envelope's integral (iEMG) as a metric of the total area of the envelopes. EMG onset and offset, required for the calculation of iEMG, was detected using the Teager-Kaiser Energy Operator (TKEO) (LI; ARUIN, 2005). Whenever the automatic detection failed, manual detection with the aid of the TKEO generated threshold was performed. All the iEMG values were time-normalized to remove the effect of different contraction durations (HILDENBRAND; NOBLE, 2004).

Statistical analysis included the assessment of data distribution according to the Shapiro-Wilk test, and the calculation of descriptive statistics. In order to evaluate the differences between the CE and TE across the sub-maximal loads, an ANOVA for repeated measures test was carried out. All statistic procedures were performed using SPSS Statistics 18 (IBM, USA), with a significance level of 95% ($p < 0.05$) and following the Greenhouse-Geisser



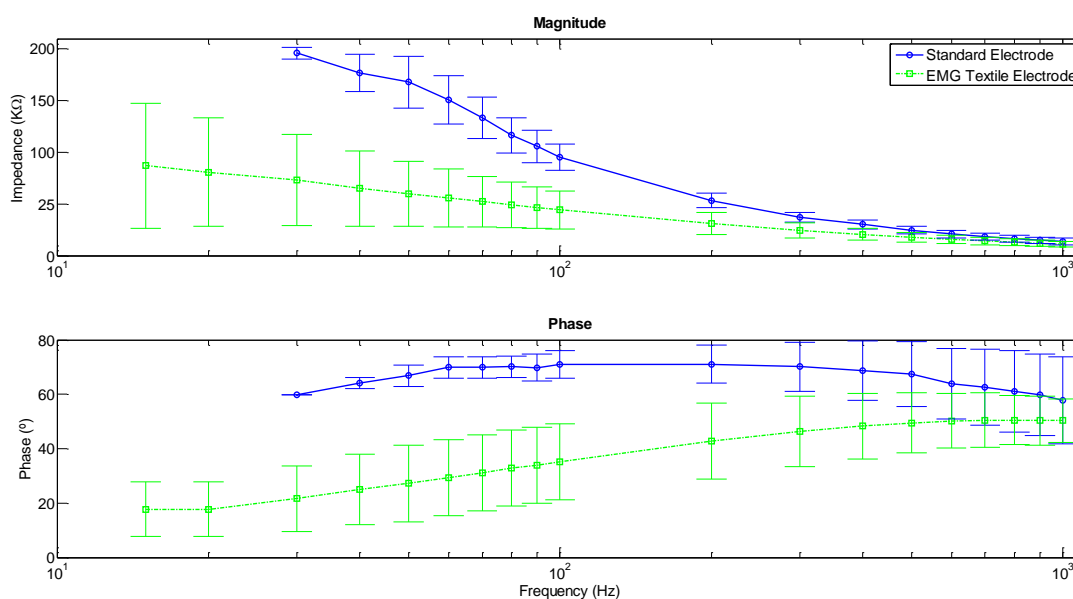
line to guarantee data homogeneity. Agreement between the CE and TE measurements of the pEMG and the iEMG were assessed using Bland-Altman diagrams.

RESULTS

Both types of electrodes present an impedance decrease with the increase of frequency, as depicted in Figure 2. However, CE have a higher impedance, especially at lower frequencies where it achieves 200 k Ω . At higher frequencies, CE reduces its amplitude to 13.6 k Ω after showing a sharper transition at 100 Hz. The TE presents lower impedance across all the tested frequencies, with an almost linear decreasing trend.

Similarly, electrodes present distinct impedance phase differences across frequencies. TE introduces a lower phase shift than CE, but it increases almost linearly with the signal frequency. On the other hand, CE shows a decreasing shift at 500 Hz.

Figure 2 – Measured impedance magnitude and phase from the skin-electrode interface



Source: authors's owns.

The ratio of signal to noise measured for each electrode is summarized in Table 1. On average, the difference between electrodes is only one decibel. However, the standard deviation and the difference between the maximum and minimum of each type of electrodes reveal that the TE have a higher range than the CE.



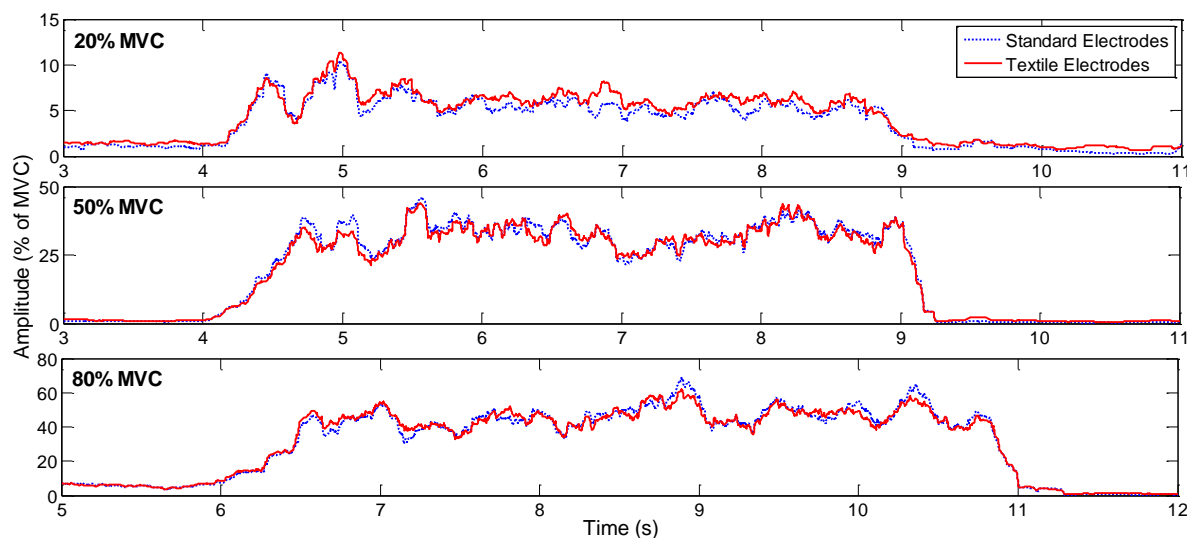
Table 1 – Average signal-to-noise ratio of the standard and textile electrodes during maximum voluntary contractions.

	Conventional (dB)	Textile (dB)	Difference (dB)
Mean	23.4	22.4	1.0
Standard Deviation	4.6	5.6	1.0
Minimum	16.9	13.5	3.4
Maximum	36.3	37.5	1.2

Source: authors's owns.

The shape of the envelopes may provide some insight on the ability of the TE to measure the same EMG information than CE. The envelopes corresponding to the sub-maximal loads are depicted in Figure 3. Despite some differences during the resting period, at 20% MVC the envelope of both signals seems to present an almost complete overlapping (Figure 3). This matching of features and dynamics seems to improve with load.

Figure 3 – Overlapping of the RMS envelopes of the EMG signals recorded with the conventional and textile electrodes at different sub-maximal loads.



Source: authors's owns.

However, the results of the RMSE reveal that the differences between signals are, on average, lower at 20% and 80% MVC, and a high variation does occur as reported by the mean and standard deviation in Table 2.

When looking at the pEMG, the CE reveals a higher value along all loads. This higher activation leads to a higher difference with TE with the increase in the load. The iEMG,



representing the area of the envelope, presents a similar trend, with CE showing higher values when compared with TE.

Despite these apparent dissimilarities, the ANOVA analysis found no differences between measurements ($p=0.48$) or between the measurements and electrodes ($p=0.75$), which indicates that both types of electrodes can measure similar values of pEMG. Finally, no differences were found between electrodes measurements at different sub-maximal loads ($p=0.97$), which indicates that for the same load, both electrodes returned similar measurements.

Table 2 – Comparison of envelope morphological parameter between conventional and textile electrodes, expressed as mean (standard deviation) and range (minimum- maximum)

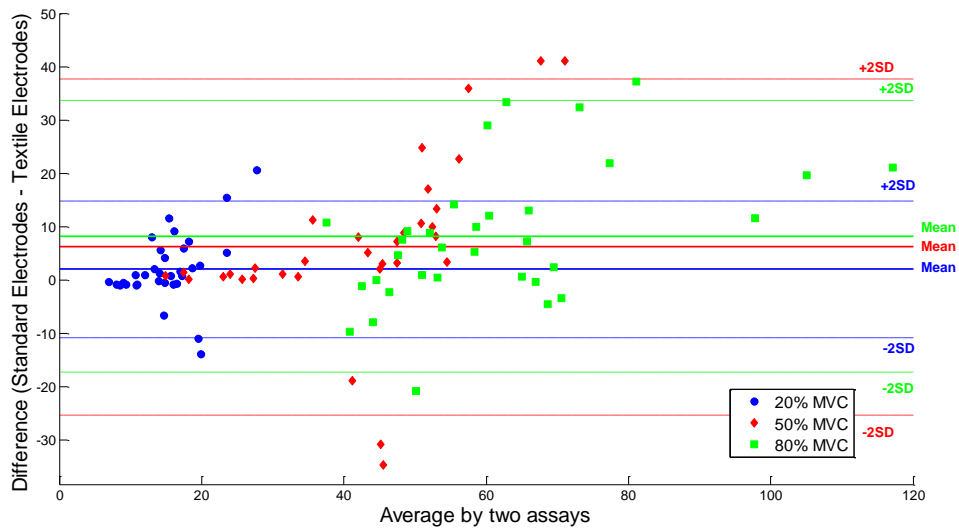
Load	Electrode	RMSE		pEMG	iEMG
		(%MVC)		(%MVC)	(%MVC)
		Mean (SD)	Range	Mean (SD)	Mean (SD)
20% MVC	Conventional			16.10 (6.99)	9.66 (3.97)
	Textile	3.6 (3.2)	0.8 – 11.5	14.16 (4.74)	8.77 (3.23)
	Difference			1.93 (2.25)	0.89 (0.74)
50% MVC	Conventional			41.22 (23.04)	29.01 (15.73)
	Textile	9.3 (9.5)	1.2 – 31.5	36.42 (16.61)	24.46 (8.98)
	Difference			4.79 (6.44)	4.55 (6.75)
80% MVC	Conventional			60.33 (29.03)	38.39 (14.14)
	Textile	7.9 (4.9)	1.9 – 22.1	52.80 (23.55)	35.09 (11.29)
	Difference			7.53 (5.48)	3.30 (2.85)

Source: authors's owns.

The agreements between pEMG measurements were analyzed with the Bland-Altman plot and the results are depicted in Figure 4. The positive bias of the pEMG difference shows that the CE generally have higher peak amplitude, which becomes higher with the increase of the sub maximal load. Just a few data points can be found outside the limit of agreement established by the ± 2 SD interval, which account for 10% (at 20% MVC), 13.3% (at 50% MVC) and 6.7% (at 80% MVC) of the total data points.



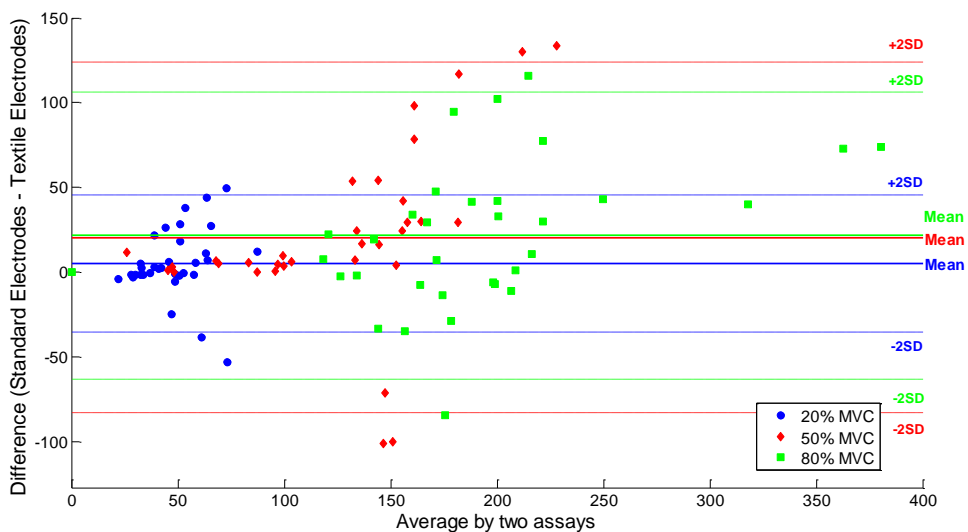
Figure 4 – Bland-Altman diagram of the pEMG values for each load



Source: authors's owns.

Similar agreement analysis was performed with the iEMG, and the results are reported in Figure 6. The difference bias shows that the CE tend to have higher values than the TE, with the bias increasing with load. However, the bias between 50% and 80% is smaller indicating that at these levels of activation, the iEMG are more similar.

Figure 5 – Bland-Altman diagram of the iEMG value for each load



Source: authors's owns.



DISCUSSION

The skin-electrode impedance is very different between electrodes. The CE has higher impedance and phase shift, which may cause differences to the measured signal across the frequency spectrum. Although the TE and CE have similar impedance at high frequencies, the EMG physiological signals usually have relevant information only up to 500Hz. Therefore, the TE has a superior performance by inducing lower impedance and phase shift to the EMG signals in this range.

Since the electrical noise affects the amplitude of both the isoelectric lines and the myoelectric signal in the same manner, the fact that CE and TE present similar amplitude seems to indicate they also share the same susceptibility to noise. However, the evaluation performed by the SNR calculation of both signals revealed smaller average differences, when compared with the results obtained by authors like Chan and Lemaire (2010), and Laferriere, Lemaire and Chan (2011), that also compared non-conventional electrodes with CE for EMG applications. These results indicate that both electrodes have a similar susceptibility to the power-line noise at 50 Hz and its harmonics.

The morphological analysis of the signal was intended to observe if there were any significant pattern differences between the signals measured by the CE and TE. The need of such assessment is related with the influence of the electrode's different manufacturing process, shape, and the simultaneous placement over the muscle. According to Hermens et al. (2000), the difference in shape is not an issue as long as the total area is similar. This did in fact occur since the TE and CE had an identical sensory area. Therefore, the only concern that remained was the difference in the electrode placement. The biceps brachii muscle was chosen because it was large enough to accommodate two pairs of electrodes over its belly. When qualitatively compared, signals recorded with both types of electrodes did not show any decisive difference between their overall patterns. Although there are differences between the CE and TE envelopes, they seem to coincide in most of their features, has revealed by the matching of several peaks. The RMSE between the envelopes of each type of electrode presented small differences (average <10 %MVC) over the entire signal length which enabled the TE to measure patterns identical to CE.

The TE seems to be capable of measuring pEMG with smaller amplitudes than the CE. This can be explained by the practical impossibility of placing each pair of electrodes on



the exact same muscle location. The small distance between each pair of electrodes, and the fact that TE was placed distal to CE, was enough to cause amplitude differences between the recorded signals (BECK et al., 2008; RAINOLDI et al., 2000). Results support this statement since the CE presented higher amplitudes along all the sub-maximal loads. Despite the lower pEMG mean, the small standard deviation of TE shows they have a smaller variability when measuring this parameter.

The iEMG results put into evidence that CE have a greater area under its envelope at all the sub-maximal loads. This is in accordance with the pEMG results, and the hypothesis that the lower amplitude is related to the distal placement of TE. The Bland-Altman diagram distribution is very similar between pEMG and iEMG, denoting a close relationship between these two variables. The small difference in terms of mean and standard deviation, and the identical range established by the minimum and maximum values reveal that the CE and TE measure identical EMG signals. The results provided by the ANOVA statistical analysis ensured that the measurements performed with both types of electrodes at different sub-maximal loads do not have differences.

CONCLUSION

The EMG signals recorded with a TE electrode were characterized in terms of electrical and morphological characteristics and compared with a CE.

Morphological analysis revealed that the textile electrodes are capable of measuring signals with similar patterns but with smaller amplitude. However, caution is advised as this may be the result of different electrodes placement locations.

The overall noise content was not substantially different between electrodes and did not compromise the ability to interpret the textile electrode's signals. Further studies should be performed to evaluate the frequency content of the signals, which may vary due the different impedance and phase shifts found. This may be particularly important in during dynamic activities.

From the results obtained in this study, we can conclude that textile electrodes have a good measuring performance during isometric conditions, allowing the measurement of EMG signals with morphological features identical to the standard electrodes. Thereby, the textile electrodes may be used as a reliable alternative to the conventional silver-chloride



electrodes in EMG measurement applications without any trade-off of signal quality or pattern alterations.

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