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Feasibility Testing of Hydrophobic Carbon Electrodes for Acquisition of Underwater Surface Electromyography Data

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Abstract—Underwater surface electromyography (sEMG) signals are especially of interest for rehabilitation and sports medicine applications. Silver/silver chloride (Ag/AgCl) hydrogel electrodes, although the gold standard for sEMG data collection, require waterproofing for underwater applications. Having to apply waterproof tape over electrodes impedes the deployment of sEMG in immersed conditions. As a better alternative for underwater applications, we have developed carbon black/polydimethylsiloxane (CB/PDMS) electrodes for collecting sEMG signals under water. We recruited twenty subjects to collect simultaneous recordings of sEMG signals using Ag/AgCl and CB/PDMS electrodes on biceps brachii, triceps brachii, and tibial anterior muscles. The Ag/AgCL electrodes were covered in waterproof tape, and the CB/PDMS electrodes were not. We found no differences in sEMG signal amplitudes between both sensors, for the three muscles. Moderate mean correlation between Ag/AgCl and CB/PDMS electrodes was found on the linear envelopes (≥ 0.7); correlation was higher for power spectral densities (≥ 0.84). Ag/AgCl electrodes performed better in response to noise, whilst the CB/PDMS electrodes were more sensitive to myoelectric activity in triceps and tibialis, and exhibited better response to motion artifacts in the measurements on the triceps and tibialis. Results suggest that sEMG signal collection is possible under water using CB/PDMS electrodes without requiring any waterproof or adhesive tape.

Keywords—Carbon black-PDMS, Electrodes, sEMG data, Waterproof.

INTRODUCTION

Silver/silver chloride (Ag/AgCl) hydrogel electrodes are the gold standard for surface electromyogram (sEMG) signal collection for dry conditions. Collection of sEMG data under water is also carried out with Ag/AgCl electrodes, especially for rehabilitation and sports medicine applications.[4,8,20,25,26] However, to collect reliable sEMG data under water, Ag/AgCl electrodes require insulation from water penetration. Typically, the waterproofing procedure requires covering the electrodes with a water-resistant adhesive film, or other even more cumbersome or abrasive procedure involving sealing electrodes with putty, adhesive bandage or foam pads.[2,4,6,9,20,24,38,39] These complicated procedures impede the extensive deployment of sEMG in immersed conditions. More importantly, sealing electrodes with adhesive tape from water penetration can causes skin irritation and abrasion due to skin peeling off when the tape is removed.

Hence, electrodes that are easily-applied and not irritating to skin are very much needed, to collect sEMG under water. Recently, we developed carbon black/polydimethylsiloxane (CB/PDMS) electrodes for collecting electrocardiogram (ECG) signals under water.[37] CB/PDMS electrodes, which were found to be fully functional not only in fresh water, but also in chlorinated and salt water,[31] exhibited good response, even in the presence of motion artifacts.[37] We hypothesize that the same CB/PDMS electrodes which we demonstrated were feasible for ECG can also be used for sEMG data collection in underwater conditions.

Electrical activity of muscles can be detected using EMG. EMG can be collected using indwelling electrodes or surface electrodes, in which case it is termed surface EMG (sEMG). Underwater EMG recording is widely used.[15,21,32,35,40] For example, rehabilitation treatments in exercise pools, such as using an aquatic treadmill,[20] are often recommended due to the advantages of hydrostatic forces and drag unique to

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the aquatic medium. EMG recordings are used to examine muscular activation during movements or to monitor the effects of the aquatic treatments. Furthermore, EMG signals can be used to assess the effect of microgravity on the neuromuscular system, or to validate ergonomic characteristics of the International Space Station with the reconstruction of part of it in a water pool, using the dry water immersion model.[22] Almost all underwater EMG studies are focused on the signal amplitude to infer information about the muscular activity level[11,13,15,21,35,36] and the movement phases.[10,32,40] Although the use of indwelling needle electrodes obviates the need for waterproofing,[32,33] the use of sEMG provides muscle activity data that are more representative of the total muscle recruitment activity than the more localized results obtained from indwelling needle EMG. Moreover, sEMG procedures can be used during movements that may not be ideal for indwelling electrodes. However, surface electrodes require insulation from water contact between the electrode, the, and wire-lead connections.

The standard gel Ag/AgCl electrodes, which exhibit good adhesion to skin after adequate skin preparation, are the universal option for biopotential signal collection in clinical and research applications, including sEMG.[7] However, applications for gel Ag/AgCl electrodes are limited by skin irritations, bacterial growth especially for long-term recordings, gel dehydration over time, and signal degradation with sweat.[19] Furthermore, Ag/AgCl electrodes have limited shelf life, which complicates inventory management.

Electrodes for sEMG data collection that function even when water infiltrates the electrode–skin interface do not exist. Previously, we presented CB/PDMS electrodes for underwater monitoring, originally tested for ECG measurements.[37] As both ECG and sEMG electrodes are designed to capture biopotential signals, CB/PDMS electrodes, which were shown to be fully functional even in water immersion, should also be applicable for sEMG data collection. Note, however, that the nature and characteristics of EMG signals are different than those of ECG, hence, the aim of this study was to evaluate the feasibility of CB/PDMS electrodes for underwater sEMG data collection, and to compare their performance to the gold-standard Ag/AgCl electrodes.

MATERIALS AND METHODS

Twenty healthy volunteers (12 males, 8 female) of ages 24 ± 8.3 years old (mean ± SD), weight 68 ± 10.7 kg, and height 171.3 ± 9 cm, were enrolled in this study.

Fabrication of CB/PDMS Electrodes

Fabrication of CB/PDMS electrodes has been described in our previous studies.[31,37] An abbreviated description of the fabrication procedure for the CB/PDMS electrodes with an embedded copper mesh layer is summarized here. The 3D-printed Acrylonitrile Butadiene Styrene (ABS) cavity molds (Objet350 Connex, Stratasys, Eden Prairie, MN, USA) were filled with the conductive CB/PDMS composite and leveled so that no excess material remained. Then, a highly conductive layer was affixed on the CB/PDMS mix to allow signal acquisition *via* the monitoring device. An insulated and waterproofed wire was soldered to the material of the highly conductive layer and used as a connector to an ECG monitoring device. A PDMS and curing agent mixture was then used to encapsulate the exposed surface with embedded copper mesh. All components were degassed for 15 min in a vacuum chamber. The fasteners were soldered to the exposed end of the wire extending from the electrode. The completed electrode assembly was then placed in a curing oven at 75 °C for 3 h. Finally, after 3 h the molds were removed from the curing oven and subsequently the electrodes were also removed from the cavity molds. For this study, as CB/PDMS electrodes were fabricated specifically for sEMG data collection, the dimensions of a single electrode were 7/8 in × 1−/2 in. In addition, a layer of PDMS was coated on top of the electrodes, to reduce noise contamination observed in the original CB/PDMS electrodes in the spectral band of sEMG signals.

Electrode–Skin Contact Impedance Measurements

Electrode–skin impedance was analyzed for Ag/AgCl (GS28 Solid Gel—Foam Electrode—7/8 in × 1−/2 in, Bio-medical instruments, Clinton Township, Michigan, USA) and CB/PDMS electrodes. An IM3570 (LCR meter and impedance analyzer with range from 100 to 100 MΩ, 4 Hz to 5 MHz, 5 digits setting resolution, minimum resolution 10 mHz, Hioki, Nagano, Japan) was used for impedance measurements. All measurements were collected on the same subject. The subject's skin was cleaned before each measurement by wiping with a 2%-alcohol infused cotton pad; the alcohol was fully evaporated before applying the electrodes. Each pair of identical (Ag/AgCl or CB/PDMS) electrodes were mounted on the left forearm, one on the palm side of the wrist, and the second 5 cm apart from the first but situated towards the elbow. The signal voltage amplitude was set to 1 V and the frequency range varied from 4 to 3 kHz. Measurements were performed in a single day to ensure constant skin properties.

sEMG Data Collection Protocol

We adapted a protocol used in a previous study[34] to underwater conditions. To ensure accurate comparison between the two types of electrodes, we recorded simultaneous measurements. To accomplish this, we placed CB/PDMS and Ag/AgCl electrodes side-by-side (Fig. 1). CB/PDMS and Ag/AgCl electrodes were placed in a lateral position (left or right on the same muscle) that alternated from subject to subject, to eliminate any bias from being only on one side. We mounted Ag/AgCl electrodes first, then waterproofed them with Tegaderm film (3 M Company, Maplewood, Minnesota, USA) prior to placing the CB/PDMS electrodes on the arm and wrapping a Velcro fabric strap around it to keep the CB/PDMS electrodes in place (Fig. 1, left). Each pair of electrodes of the same type were placed in the longitudinal direction on the muscles about 3 mm apart from each other to minimize cross talk, and both types of electrodes were separated transversely by no more than ~ 1 cm (Fig. 1). Electrodes of the same size were used on the three muscles included in this study, for both types of electrodes.

We acquired sEMG signals using a Dual Bio Amp (ADInstruments) and digitized them at a sampling frequency of 2 kHz. We recorded sEMG measurements on three muscles: the biceps brachii, triceps brachii (long head), and tibialis anterior. The sampling frequency was selected to meet the requirements for the biceps brachii and the tibials anterior.[3,18,30] For each muscle, we began the recording with 5 s of data while the muscle was in relaxation, then the subject transitioned to a contraction stage, which consisted of isometric contraction for 6 s followed by relaxation for 5 s. Between the different stages there were 2 s of transition time. The time sequence can be summarized as: 5 s relaxation, 2 s to move to contraction stage, 6 s of maintaining contraction, 2 s to move to relaxation stage, and 5 s of relaxation. We followed the same time sequence for sEMG signal recording on all three muscles. Subjects practiced the maneuvers prior to every test until they felt comfortable with the procedure.

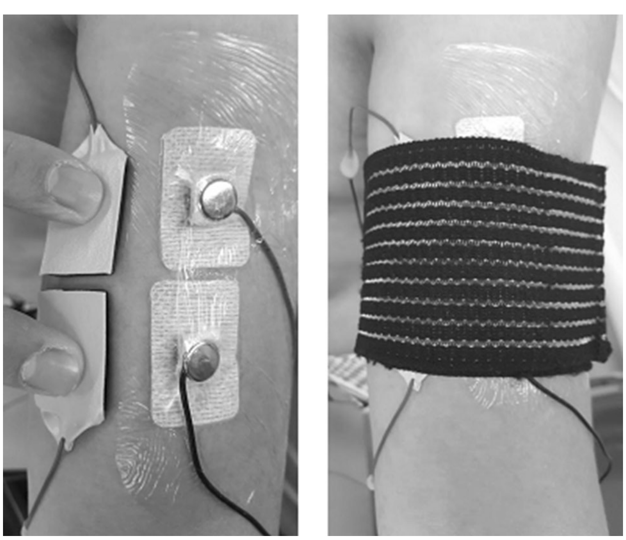

FIGURE 1. Left: CB/PDMS and Ag/AgCl electrodes on a subject's biceps. Notice that Ag/AgCl electrodes shown on the right side of the left figure are waterproofed with Tegaderm ®. Right: Velcro strap to keep electrodes in place.

Figure 2 shows where we placed the electrodes on each muscle.[17] The electrodes were placed with the subjects resting. We recorded sEMG measurements of the three types of muscles while the subjects performed three muscle contraction maneuvers during the experiment, one for each muscle. These specific muscles were chosen based on their different sizes, which allowed us to test the electrodes at different levels of amplitude, and the feasibility to perform contraction maneuvers inside the water bucket. Before performing every test, we made sure that the location where the electrodes were placed was hairless and had been wiped with alcohol and allowed to dry.

The contraction maneuver was different depending on the muscle, to ensure appropriate contraction. For the biceps, the subjects lifted a weight of 6 lbs. (2.72 kg) from the bottom of the bucket, bringing the elbow to a 90° angle, with the forearm in supination. To ensure proper contraction of the triceps, subjects pushed down a resistance band with the palms facing down (a triceps pushdown), bringing the elbow to a 180° angle. The tibialis anterior muscle was contracted by having the subject lift the sole of their foot off the bottom of the bucket, while keeping the heel on the floor, without extension of the great toe (Fig. 2). This study was carried out in accordance with the recommendations of the Institutional Review Board of The University of Connecticut. All subjects gave written informed consent in accordance with the Declaration of Helsinki. The protocol was approved by the Institutional Review Board of The University of Connecticut.

sEMG Data Processing

We processed sEMG signals offline to quantify their quality and to compare the performance of CB/PDMS electrodes to the taped Ag/AgCl electrodes. First, to test the similarity in signal dynamics between CB/PDMS and Ag/AgCl sEMG data in the time and frequency domains, the Pearson's correlation coefficient was computed in the time and frequency domains to test interchangeability between CB/PDMS and water-insulated Ag/AgCl electrodes. Furthermore, in the time domain we computed the linear envelope, ampli-

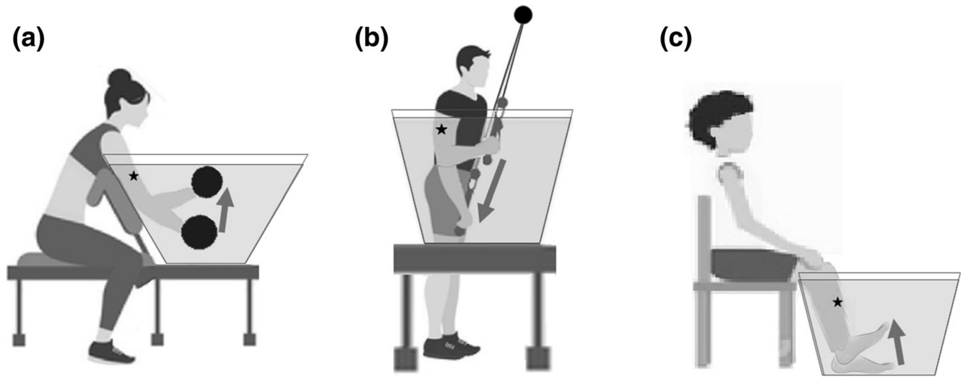

FIGURE 2. Experimental protocol. Stars represent approximate electrode placement locations in: (a) biceps brachii, (b) triceps brachii (long head), and (c) tibialis anterior.

tude, and root mean square (RMS) envelope. In the frequency domain, the signal-to-noise (SN) ratio, signal-to-motion (SM) ratio, the maximum-to-minimum drop in power (DP ratio) and the spectral deformation (Ω ratio) were computed.

For frequency domain analysis, the power spectral density (PSD) of each sEMG signal was calculated using Welch's periodogram method with 50% data overlap. A Blackman window (length of 256 data points) was applied to each segment and we calculated the fast Fourier transform (FFT) for each windowed segment. Finally, we averaged the power spectra of the segments. An FFT segment size of 1024 data points was used. The procedures to compute all indices are described below.

Linear Envelope

One rectifies sEMG signals by taking their absolute value, low-pass filters at 10 Hz, and down-samples to 41.66 Hz (a rate that is closer to motion frequencies) to get a linear envelope. The resulting envelope is an estimate of the standard deviation of the sEMG signal, which is in turn a measure of the muscles' power. The Pearson's correlation between CB/PDMS and Ag/AgCl electrodes' sEMG envelopes was computed, for each muscle type, to test the similarity between the two simultaneously-acquired signals. Correlation provides an index of similarity, independent of the amplitude of the signals which were collected with the two types of electrodes side by side.

Amplitude

The mean value of the linear envelope is computed as an amplitude estimation of sEMG signals. This index was computed for relaxation and contraction stages, to evaluate the statistical difference in amplitudes between the signals obtained using CB/PDMS and Ag/AgCl electrodes.

RMS Envelope

We divided the sEMG signals into multiple windows of 25 ms[12] and computed the RMS values from the signals before rectification, as the values have both negative and positive values. As with the linear envelope, the Pearson's correlation between CB/PDMS and Ag/AgCl electrodes' sEMG RMS values were computed for each muscle.

SN Ratio

This index considers noisy disturbances in the high-frequency range of the PSD.[1] For the SN ratio calculation, we assumed that noise had a constant power density over the frequency region of interest in sEMG recordings and that no muscular activity-related power was present above 800 Hz (the upper 20% of the frequency range). Thus, first, the power for the frequency range above 800 Hz was calculated. The predicted total power of the noise is this power summed over the whole frequency range. The SN ratio was then calculated as the ratio of the total sEMG power to the total power of the noise.

SM Ratio

For this study, motion artifacts are defined as low-frequency fluctuations of the signal induced by mechanical alteration of the electrode–skin interface. Use of the SM ratio is based mainly on two assumptions: (1) the frequency of motion-induced artifacts of the signal stays well below 20 Hz, and (2) the shape of the non-contaminated sEMG power spectrum is fairly linear between 0 and 20 Hz.[41] Consequently, the motion artifacts' spectral power will be mixed in with the true signal dynamics at frequencies between 0 to 20 Hz. Per Sinderby *et al.*,[41] the motion artifacts' power (the grey area in Fig. 3) can be reasonably estimated by summing the PSD area below 20 Hz that exceeds a straight line between the origin and the highest mean power density. The highest mean power density (the red dot in the averaged spectral plot of Fig. 3) was defined as the largest mean spectral value within a window length of 25.4 Hz, in the range 35–500 Hz. Finally, the sum of the area under the PSD for all frequencies divided by the motion artifact power was computed to obtain the SM ratio.

DP Ratio

One obtains the DP ratio by computing the quotient between the highest and lowest mean PSD values. The mean PSD is obtained by averaging a spectral window length of 25.4 Hz (13 consecutive points). The DP ratio is an indicator of whether the spectral frequency contents of interest are adequately peaked. The DP ratio is sensitive to the signal's amplitude, and can detect the absence of sEMG activity. It is not sensitive to power below 35 Hz (in contrast to the SN ratio) and will not provide falsely high values because of the power induced by motion artifacts. A higher DP ratio is desirable.

Ω Ratio

The spectral deformation is computed in terms of spectral moments, as follows:

$$\Omega = (M_2/M_0)^{\frac{1}{2}}/(M_1/M_0),$$

where

$$M_\mathrm{n} = \sum_{i=0}^{i_{\max}} \text{power density}_i \cdot \text{frequency}_\mathrm{i}^\mathrm{n}$$

The Ω ratio is sensitive to changes in symmetry and peaking of the PSD and to additive disturbances in the high- and low-frequency regions.[1] This index identifies all dynamics of spectral changes except those caused by pure translations along the frequency axis. The feature is also sensitive to an excess of low-frequency power. A lower Ω is desirable.

The SN, SM and DP ratios are presented in decibels, and the Ω ratio is unitless. These four indices obtained for CB/PDMS and waterproofed (taped) Ag/AgCl electrodes were compared, by testing for statistically significant differences, to examine whether there is an electrode media that collects the signal with lower noise power, lower motion-artifact corruption, more sensitivity to EMG activity, and lower distortion. This battery of indices has been used before to assess the quality of myoelectric signals.[5,27,34] The normality of the quality measures was tested using the one-sample Kolmogorov–Smirnov test.[23,29,42] To test the null

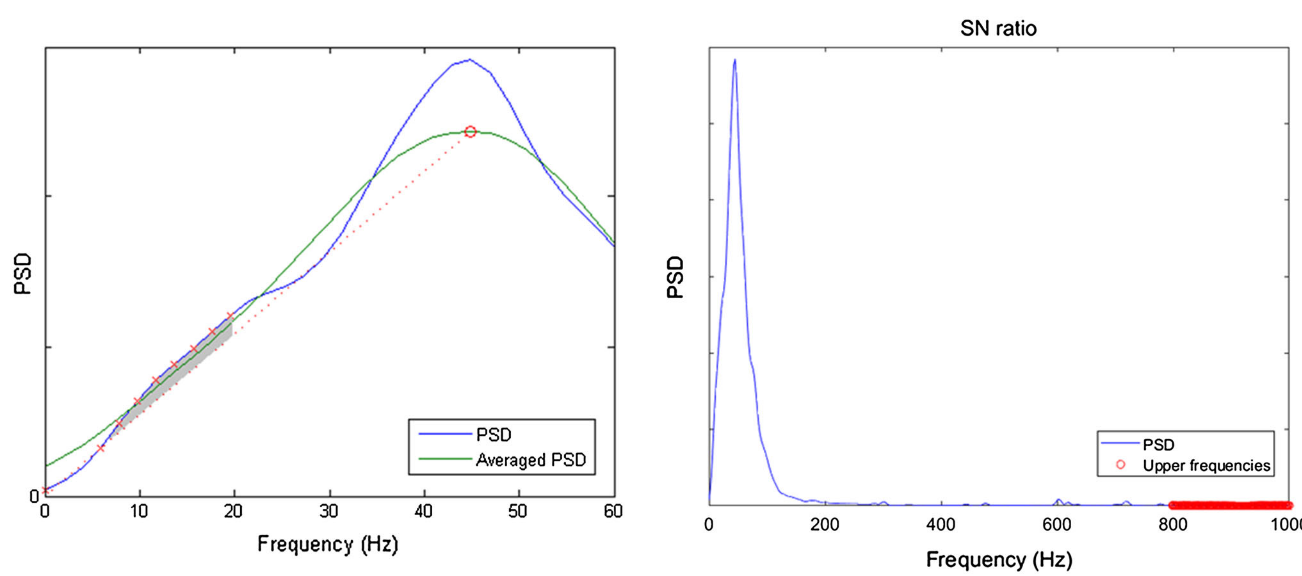

FIGURE 3. Illustration of SM ratio (left) and SN ratio (right) estimation.

hypothesis that the above-mentioned indices are equal for CB/PDMS and Ag/AgCl electrodes, we used the t test if measures were normally distributed, or a two-sided Wilcoxon rank sum test if non-normality was found.[16] A p value < 0.05 was considered significant.

RESULTS

Results for electrode–skin contact measurements are presented in Fig. 4. Showed values are the result of averaging 20 consecutive measurements. The impedance of CB/PDMS electrodes is higher compared to Ag/AgCl throughout the range 4 Hz to 3 kHz. This is in agreement with previous studies,[31,37] considering the size of the electrodes used in this study. Figure 5 shows representative sEMG signals acquired using Ag/AgCl and CB/PDMS electrodes on a specific subject's biceps. Despite the possible influence of the location in the amplitude of the signal, both electrodes show sEMG signals with similar amplitudes. Table 1 contains the results for our amplitude comparison between Ag/AgCl and CB/PDMS electrodes. Amplitude measurements were found to be non-normally distributed. No significant differences in amplitude were found between the two media, for relaxation and contraction stages, in biceps, triceps and tibialis sEMG measurements. Furthermore, a Bland–Altman plot (Fig. 6) shows the small bias (-11.3 μV) and low variance (4.6×10^{-3}) between the amplitude measurements obtained using the two media.

The linear envelope and the RMS value envelope were moderately correlated between Ag/AgCl and CB/PDMS sEMG signals. For biceps, triceps and tibialis, the average correlation was higher than 0.7. In the frequency domain, the PSD of signals obtained using the two types of electrodes were highly correlated (≥ 0.84 on average).

Table 2 includes the frequency-domain indices for quality assessment of sEMG signals. Non-normality was found in SN and SM ratios, while DP and Ω ratios were normally distributed. The SN ratio was significantly different between the two media for the three types of muscles tested in this study. Ag/AgCl electrodes showed significantly higher SN ratio, compared to CB/PDMS. The SM ratio was statistically higher in triceps and tibialis for CB/PDMS electrodes, compared to Ag/AgCl. The DP ratio was significantly higher for CB/PDMS electrodes in triceps and tibialis, compared to Ag/AgCl electrodes. Finally, the Ω ratio was significantly higher for the CB/PDMS electrodes in biceps and tibialis.

DISCUSSION

We were successfully able to acquire sEMG signals under water, using dry electrodes that do not require insulation from water penetration. This is a great improvement from the only good current option, which is to place waterproof tape over standard gel electrodes, irritating the skin. Despite higher electrode–skin impedance, our CB/PDMS electrodes were able to collect sEMG signals with no reduction in amplitude when compared to the gold standard Ag/AgCl hydrogel electrodes, which had to be fully insulated using waterproof tape. Although noise more significantly affected the signals acquired with CB/PDMS electrodes, and higher spectral distortion was detected (Ω ratio), they exhibited a better response to motion

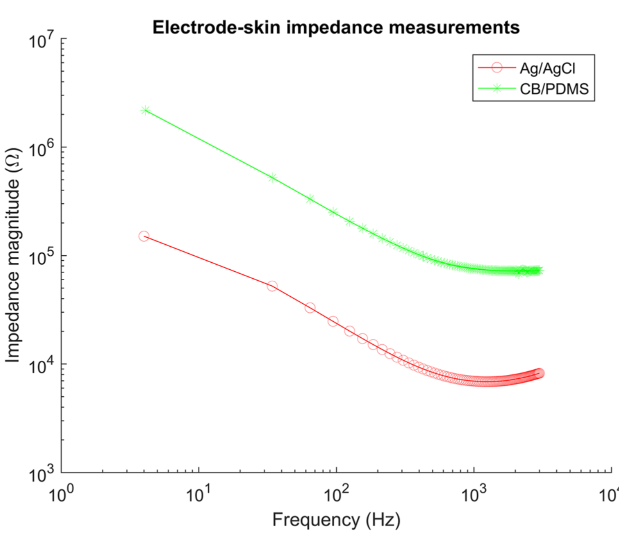

FIGURE 4. Electrode-skin contact impedance measurements for Ag/AgCl and CB/PDMS electrodes.

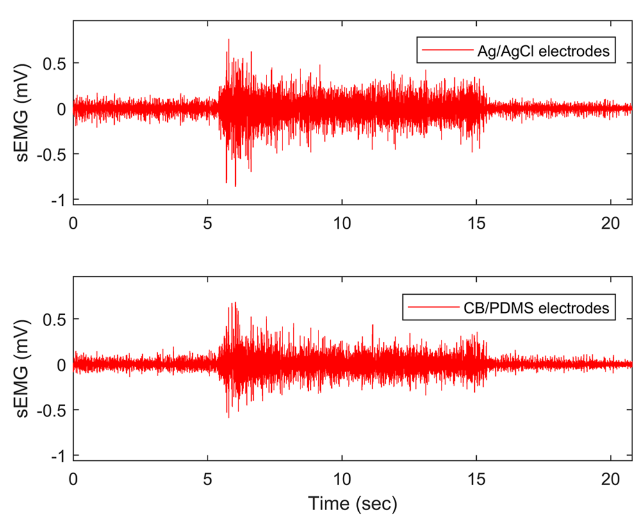

FIGURE 5. sEMG signal acquired under water using Ag/AgCl (top) and CB/PDMS electrodes (bottom), for a given subject's biceps.

TABLE 1. Amplitude of sEMG signals collected using Ag/AgCl and CB/PDMS electrodes.

	Biceps		Triceps		Tibialis	
	Relaxation	Contraction	Relaxation	Contraction	Relaxation	Contraction
Ag/AgCl	30.1 ± 37.8	80.5 ± 60.5	37.1 ± 70.9	55.9 ± 104	27.4 ± 22.5	55.3 ± 34.1
CB/PDMS	30.9 ± 28.3	91.5 ± 35.3	38.9 ± 26	57.5 ± 33.1	34.9 ± 33.3	76.4 ± 49.2
Envelope correlation	0.73 ± 0.23		0.72 ± 0.28		0.7 ± 0.26	
RMS correlation	0.75 ± 0.23		0.74 ± 0.28		0.73 ± 0.25	
PSD correlation	0.93 ± 0.1		0.84 ± 0.19		0.86 ± 0.14	

Values are expressed as mean ± standard deviation. Amplitude values are in mV.
CB/PDMS carbon black/polydimethylsiloxane, *Ag/AgCl* silver/silver chloride.

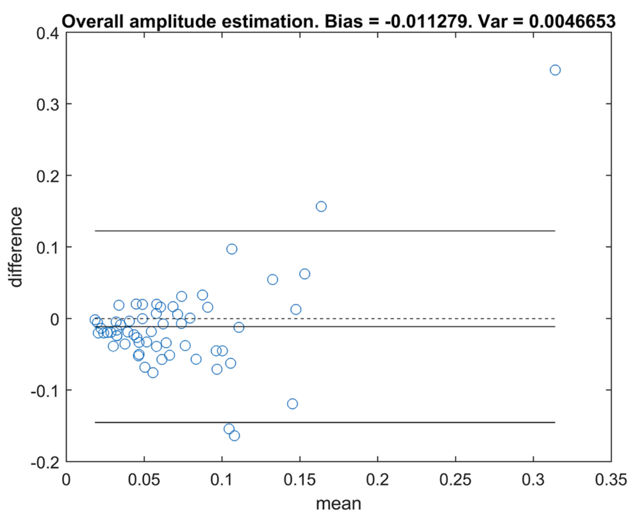

FIGURE 6. Bland–Altman plot for comparison of amplitude measures using Ag/AgCl and CB/PDMS electrodes. Bias = − 11.3 μV, variance = 4.6 × 10^{-3}.

TABLE 2. Frequency domain indices of sEMG signal quality.

	Biceps	Triceps	Tibialis
SN ratio (dB)			
Ag/AgCl	69 ± 16	72 ± 3.4	73 ± 4.1
CB/PDMS	36.9 ± 24*	24.1 ± 19*	27.7 ± 21*
SM ratio (dB)			
Ag/AgCl	1.3 ± 5.7	0.05 ± 0.17	0.02 ± 0.018
CB/PDMS	2 ± 6.4	1.3 ± 5.1*	0.6 ± 1.9*
DP ratio (dB)			
Ag/AgCl	54 ± 12	47 ± 13	42 ± 5.2
CB/PDMS	60 ± 6.2	57 ± 6.5*	55 ± 6.5*
Ω ratio (relative units)			
Ag/AgCl	4 ± 0.8	4.2 ± 0.49	4.3 ± 0.48
CB/PDMS	5.5 ± 2.3*	4.9 ± 2.1	5.1 ± 2.1

Values are expressed as mean ± standard deviation.
*Means statistically significant difference ($p < 0.05$).

artifacts, more sensitivity to the myographic activity, and were more reliably able to detect the absence of muscle activity (assessed by DP ratio), in comparison with Ag/AgCl electrodes. Note that these results were obtained with the CB/PDMS electrodes and compared to Ag/AgCl electrodes that had been sealed with waterproof tape, which provides additional protection from motion artifacts.

Results suggest that CB/PDMS electrodes are susceptible to noise corruption in the range of interest of sEMG signals (5–500 Hz), significantly affecting the SN ratio. This suggests that robust noise filtering approaches are necessary for processing the sEMG signals using CB/PDMS. However, not needing to waterproof the CB/PDMS electrodes, their reusability, their ability to collect sEMG signals continuously for a long period without any skin irritation, their sensitivity to myographic activity, and their capacity to detect the absence of muscle activity, support the usability of CB/PDMS electrodes for sEMG data acquisition in water immersion.

sEMG can be applied in orthopedics, sports medicine, rehabilitation, SCUBA diving, and other applications.[14,28] Given the considerable increase of knowledge about sEMG, many efforts have been carried out to make it accessible for subjects taking part in activities in water immersion.[2,4,6,9,20,24,38,39] Ag/AgCl electrodes are the gold standard for sEMG signal collection, but the preparation they require for functioning under water have made them impractical or cumbersome for such a task. Our reusable dry CB/PDMS electrodes do not require silver and hydrogel, and can collect bioelectrical signals in all types of water.[31] We have demonstrated for the first time that these electrodes are a possible alternative for collection of sEMG data under water under water, compared to Ag/AgCl electrodes.

In this study, we found a significant lower SN ratio for the CB/PDMS electrodes, compared to Ag/AgCl electrodes. The SN ratio is computed using raw sEMG signals, so we expect this corruption can be handled with adequate filtering. It is important to point out that Ag/AgCl electrodes were fully covered with waterproof tape throughout the test, which provided additional protection against motion and noise corruption. Correlations of the envelope and RMS value envelope, DP ratio, and spectral deformation (Ω ratio) were affected by this difference in noise corruption, as it affects the average amplitude of the sEMG signal during the relaxation and contraction stages differently, in the time domain. However, the spectral content of the sEMG signals was very similar, as the high PSD correlation values showed.

Given the relationship between muscle contraction and movement, it is especially important for the sEMG applications to avoid excessive effects of motion artifacts. Although CB/PDMS electrodes were not taped to subjects' skin by any means, and only an elastic strap was used to keep the electrodes in place, motion artifacts were a less important issue for CB/PDMS electrodes than for the Ag/AgCl electrodes (higher SM ratio). In fact, although statistically-significant differences were found only in triceps and tibialis, the average SM ratio was higher for CB/PDMS electrodes in biceps, triceps and tibialis. High variance in the SM ratio was observed for both types of electrodes.

In this work, we were able to demonstrate collection of sEMG signals under water using CB/PDMS electrodes without the use of any adhesive tape, which has been the normal practice using Ag/AgCl electrodes. It has been well-documented that adhesive tape causes skin irritation with prolonged use, and some skin tissue can tear during tape removal. Given these limitations of Ag/AgCl electrodes for underwater applications, we foresee that the use of CB/PDMS electrodes together with advanced waterproof diving instruments may open new research areas involving sEMG data collection under water.

ACKNOWLEDGMENTS

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