engineering proceedings

*Proceeding Paper*

# Types of EMG Textile Electrodes: A Comparative Study Using PCA †

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† Presented at the 3rd International Electronic Conference on Biosensors, 8–21 May 2023; Available online: https://iecb2023.sciforum.net.

**Abstract:** Identifying a suitable textile electrode that would be durable and assist in recording high quality bio-signal quality is crucial in the production of medical devices. Therefore, this study is aimed at comparing the time domain characteristics of silver-plated-polyamide-embroidered cotton (SPEC), copper-nickel-plated polyester (CNP), and stainless-steel-fabric (SSF) dry textile electromyography (EMG) electrodes through principal component analysis (PCA). The standard silver/silver chloride (Ag/AgCl) gel electrode was considered as the reference for all the test textile electrodes mentioned above. The EMG signal was measured by activation of the bicep, and tibialis anterior muscles, and the time domain features such as root mean square (RMS) voltage, average rectified value (ARV) voltage, signal to noise ratio (SNR), kurtosis, and skewness were extracted from the EMG signal. The SSF electrode outperformed CNP and SPEC electrodes. Each textile electrode exhibited signal-to-noise ratio (SNR) values comparable to that of the standard electrode. The SNR values were 24.38 dB, 17.72 dB, 15.55 dB, and 13.30 dB, for Ag/AgCl, SSF, CNP and SPEC electrodes, respectively. The performance of all the conductive textile electrodes was comparable to that of Ag/AgCl. However, the gel electrode required skin preparation and exhibited short-term stability, whereas textile electrode materials lasted long and could be used for biological signal monitoring at home without the assistance of medical professionals.

**Keywords:** stainless steel; copper-nickel-plated polyester; embroidery; EMG signal analysis; PCA

**Citation:** Etana, B.B.; Malengier, B.; Krishnamoorthy, J.; Van Langenhove, L. Types of EMG Textile Electrodes: A Comparative Study Using PCA. *Eng. Proc.* **2023**, *5*, x. https://doi.org/10.3390/xxxxx

Academic Editor(s):

Published: 8 May 2023

## 1. Introduction

Healthcare home devices have improved as a result of the advancement of medical technology and design of smart devices. In today's society, it is critical to keep track of a patient's health status at home, thanks to smart textile technology. Medical information available in the form of biological signals such as the electrocardiogram (ECG), electromyogram (EMG), and electroencephalogram (EEG) [1,2]. Electromyography (EMG) is one of the widely used diagnostic techniques to examine the electrical activity of striated muscles in terms of voltage and assess the health condition of the neuron groups that stimulate these muscles [3]. The quality of EMG signals recorded in clinical applications is directly correlated with the use of high-grade EMG electrodes. Recent developments in reusable textile sensing systems have enormous advantages as wearable devices embedded with surface EMG have opened up new opportunities in the field of rehabilitation and professional sports. This has enabled the collection of data in real-world conditions, such as home settings or athletes' outdoor practices supplanting the use of gelled electrodes [4–

7]. Evaluating the various types of conductive-textile electrodes for EMG bio-signal sensing can disclose critical information about a patient's health and can be assessed at home compared to traditional electrodes. When detecting and recording the EMG signal, there are two main concerns which influence the fidelity of the signal are to be considered. The first is the SNR, which is nothing but the ratio of the power of the signal component to the power of the noise component. In general, noise is defined as electrical signals that are not part of the desired EMG signal. The other issue is the distortion of the signal, meaning that the relative contribution of any frequency component in the EMG signal should not be altered; continuous long-term EMG monitoring with better signal quality is important for early detection and monitoring of electrical activity of striated muscles before they devolve into a series of complications. Ernest N. Kamavuako et al. compared embroidered electrodes to gel electrodes and found that embroidered EMG electrodes are a cost-effective and good substitute for the sensors used in contemporary myoelectric prosthesis [8]. However, there is insufficient research on embroidered electrodes and its comparison with conventional electrodes in terms of (1) myoelectric control of the affected muscles as a treatment, (2) study the variation of EMG signal over time caused by natural biological fluctuations, (3) the durability of the fabric electrodes for long-term use, and (4) the discomfort of fastening an electrode in the forearm at some pressure (e.g., 20 mm). The effects of ideal electrode size and the pressure exerted by the fabric worn over the electrodes on SEMG signals were investigated by Siyeon Kim, Sojung Lee, and colleagues [9]. Comparative research on the performance of electrode textile materials confined to muscle areas is difficult to find. Aljoscha Hermann et al. investigated cheap EMG pants that used various stainless steel electrodes to assess quadriceps and hamstring activity [10]. Based on varied sleeve patterns, Gozde Goncu-Berk et al. conducted comparative research on the electrical characteristics of embroidered and conventional EMG electrodes[11]. Gozde Goncu-Berk et al. claimed that their embroidery textile electrode had a lower electrical resistance and a greater SNR (dB) than conventional electrodes; however, the reproducibility and durability of the embroidery materials employed were not addressed in their studies [15]. Because of 'strict' regulations on the use of electronics parts, the functionalization of electronics incorporated into textiles must be improvised and evolved as needed. Electronic textiles have been rapidly improving in recent years, and new capabilities essential in our daily lives are projected to be made available through electronic textiles [12]. Paiva, A.; Carvalho, et al. studied the comparison and performance evaluation among different types of knitted structures combining knit loops with float and tuck loops, as well as different conductive yarns for the same knit combination, in the development of dry textile electrodes reported for the betterment of reapplication [13]. Wearable textiles is a hot topic nowadays, and the quest for the best textile electrodes with the best performance evaluation is currently on fast pace. In this study, we have devised many methods to bestow the best conductivity on common fabrics. The major strategy is to select the best performing textile electrode from three various types of conductive textiles electrodes to be incorporated into a smart garment. Better performaning textile electrode would be used as an EMG sensor for continuous signal monitoring. The optimized anatomical and physiological positioning of the electrode must be investigated because the electrode position has a significant effect on the performance evaluation. The skeletal muscle is the only one in the human body that shows voluntary movements, so we chose the biceps brachii and tibialis anterior muscles to compare the different conductive textiles EMG electrodes. The evalaution was made based on their optimal performance in EMG signal monitoring under static and dynamic conditions.

## 2. Materials and Methods

Three types of woven conductive textile electrodes: (1) silver-plated polyamide e-embroidered cotton (SPEC), (2) copper-nickel-plated polyester (CNP), and (3) stainless steel fabric (SSF) were used in this study. The SPEC electrodes were created using the Ink/Stitch program, which is an embroidery plugin for Inkscape, and then transferred to a computerized knitted textile embroidery machine (Brother670E). The electrodes were created with a stitch length of 1.5 mm, and conductive silver-colored; polyester fabric made of copper-nickel-plated polyester [14], and stainless steel electro-conductive fabric with a square resistance of 20.3 ohms and 966 GSM were prepared [15]. All the conductive fabrics used were developed with a conductive dimensional area of 20 mm × 20 mm as shown in Figure 1. A conductive snap button was added to the back of each electrode to serve as the connection point between the sensing area and the BIOPAC system connector clips. The snap was stiched into such that it did not make any direct contact with the skin. Resistance was measured from the clip to the edge of the electrode [14]. Data were collected using the BIOPAC MP360 EMG data acquisition module (BIOPAC Systems, Inc., Goleta, CA, USA).

Different physiology, sex, and other inter individual differences were considered while selecting the subjects for this study. Data collection and analysis for all electrodes were performed using the BIOPAC MP360 EMG data acquisition module (BIOPAC Systems, Inc., Goleta, CA, USA). The positive and negative terminal of the BIOPAC wires were connected to the two snap elements (Figure 1a–c) attached to the conductive textile. The ground terminal from the BIOPAC was attached to an additional elastic fabric containing a single snap button. The placement of the differential electrode (positive-negative terminal pair) and the ground were placed apart by a distance of 20 cm for Biceps (near wrist region) and 10 cm for Tibialis (below the region of interest). The interelectrode distance of the two electodes constituting the differential electrode was 2 cm. The mounted configuration of the electrodes for EMG measurement from Biceps brachii and tibialis anterior is as shown in Figure 3d. Each sEMG signal was recorded for two minutes with a sampling rate of 2000 Hz.

Performance evaluation of each electrode was based on the signal recorded under dynamic condition such as the contraction and relaxation cycles of the biceps brachii muscles while excercising with loads of 2, 4, and 6 kg. For tibialis anterior, treadmill based walking and running experiments were performed. After 5 min of the application of the electrodes, the recording was begun, so as to avoid a change in the signal quality due to sweat effect. Principal component analysis was carried out on the time domain features extracted from the EMG signal obtained from each electrode.

(**a**) (**b**) (**c**) (**d**)

**Figure 1.** EMG Textile electrodes: (**a**) Woven embroidered textile electrode; (**b**) silver+ copper platted conductive fabric; (**c**) stainless steel electro-conductive fabric; (**d**) Biopac.

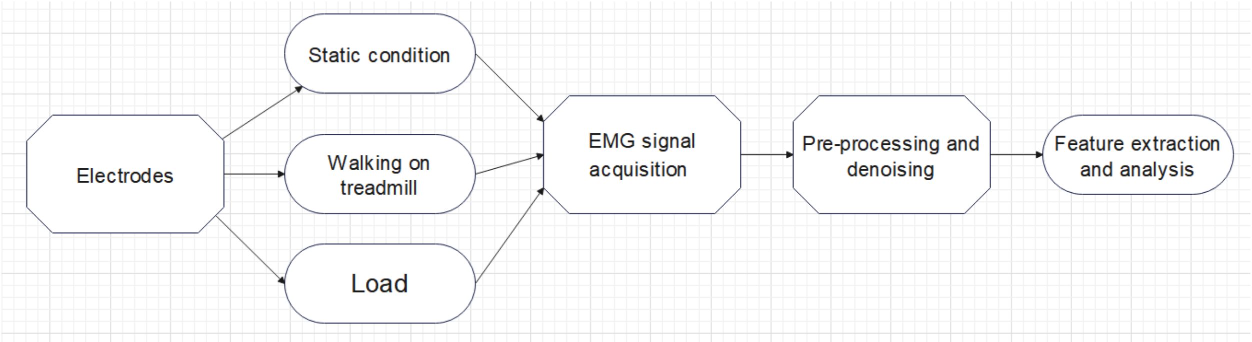


**Figure 2.** Experiment setup for wearable textile electrode-based EMG sensor.

### *2.1. Human Subject Testing*

The subjects aged from 20 to 35 years old used the Biceps and tibialis muscles. The protocol of this study was approved by Jimma university, Jimma technology Institutes of Institutional Doctoral School Review Board of materials science and engineering. All subjects were given written informed consent and provided permission for the publication of photographs for scientific and educational purposes.

### *2.2. EMG Features Extraction*

Six different types of time domain statistical features have been used in this preprocessed EMG signal and evaluated for each different type of textile electrode on the two muscle groups. These are root mean square (RMS), average rectified value (ARV), variance, standard deviation, kurtosis, and skewness. All of these features were derived from the wavelet desnoised signals [16]. The RMS feature represents the square root of the mean power of the EMG signal. Especially, this feature is related to non-fatigue muscle contraction and constant force [61].

### *2.3. Principal Component Analysis (PCA)*

A within-subjects design was selected to reduce the impact of signal variability due to different physiology, sex, and other inter individual differences. A multivariate analysis was performed a priori to determine the number of subjects necessary to see a statistically significant change. To determine differences between traditional and developed textile electrodes, we included the kurtosis feature in this study. With a single variate analysis, we would have required almost 24 subjects to see significant inter electrode differences, but by adopting multivariate analysis we can infer the same through less number of subjects as shown in the score plot (Figures 6 and 7).

## 3. Results

After assesing better signal quality through static trials experiments, several test sessions were conducted under dynamic conditions to acquire the complete experimental data. Figure 2b depicts an example of an acquired signal that depicts the different stages of the activity involving different weights or load and treadmill excercises. Signal quality for clinical electrodes and different types of textile electrodes were evaluated in confined movement scenarios. The electrical conductivity of the electrodes is one of the factors that affect the ability of the electrodes to acquire a biosignal. Figure 3 depicts that the EMG signals are often difficult to understand and interpret clinically.

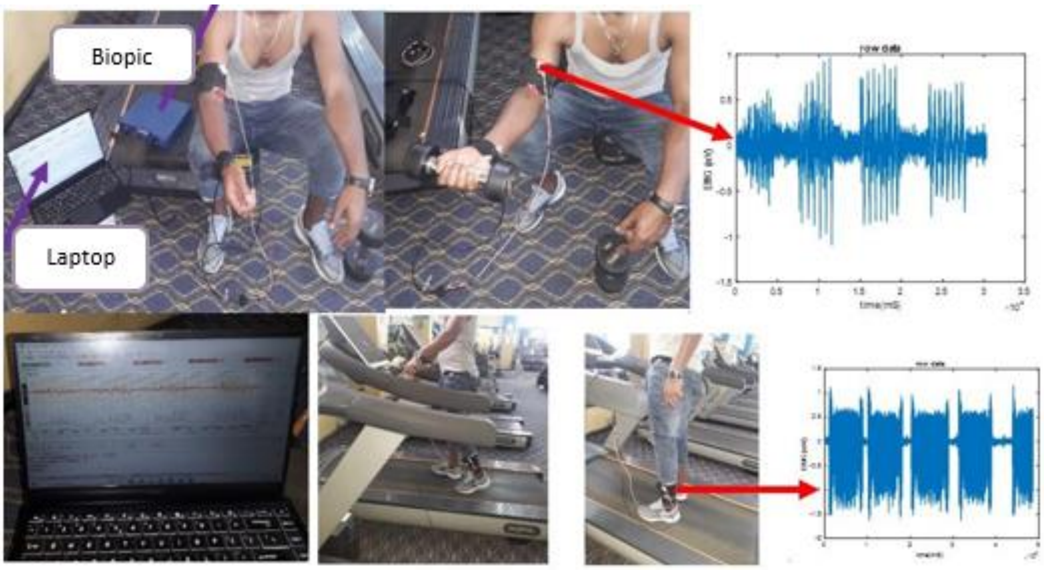


**Figure 3.** EMG measurement setup showing the fabric electrode of performed exercises with specified electrode locations to evaluate muscle response at relaxation and contraction for from biceps brachii at static (**a**); Treadmill experiment on participants where muscle activity during walking was monitored with the developed textile electrodes EMG monitor from tibialis anterior and the corresponding EMG acquisition during the dynamic condition (**b**).

Textile-based EMG electrodes were placed bilaterally on the tibialis anterior (TA), Biceps brachii muscles with minimal skin preparations. In this research work, the evaluation of different types of dry textiles electrodes such as embroidered, stainless steel, and polyester were investigated for sEMG applications. Their side-by-side comparison with clinical Ag/AgCl electrodes in terms of SNR, EMG feature extracted, and signal correlation were performed for two different muscle groups including biceps brachii and tibialis anterior. Stainless steel textile electrodes and Ag/AgCl showed high similarity for most of the extracted compared to polyester and embroidered electrode. SNR values, on the other hand, reveal that the quality of acquired signals is similar for all types of electrodes with the ratio of their SNRs for polyester textile being the minimum at 94% in comparison to over SNR of Ag/AgCl. In terms of the similarity in signal morphology, cross-correlation up to maximum of 98% and an average ~96% was achieved.

**Table 1.**

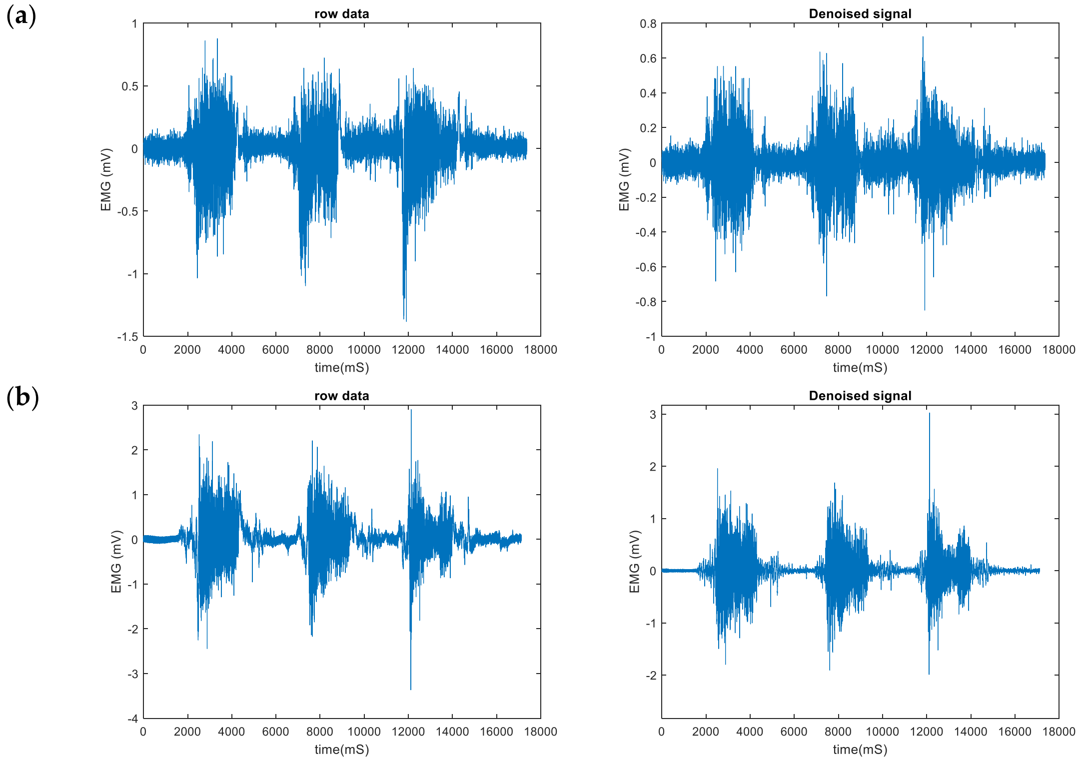
| **Muscle Group** | **Load** | **SNR (dB)** | | | | **%SNR** | | |
|---|---|---|---|---|---|---|---|---|
| | | **Ag/AgCl** | **SSF** | **SPEC** | **CNP** | **SSF** | **SPEC** | **CNP** |
| Biceps | No load | 17.72 | 15.55 | 13.3 | 24.38 | 87.75 | 75.056 | 137.58 |
| | 2 kg | 28.92 | 26.53 | 26.19 | 29.98 | 91.73 | 90.56 | 103.66 |
| | 4 kg | 19.75 | 32.56 | 33.12 | 28.92 | 164.86 | 167.69 | 167.69 |
| | 6 kg | 31.61 | 49.77 | 25.36 | 60.87 | 157.45 | 80.22 | 192.56 |
| Tibialis anterior walking (3 mph) | | 32.56 | 31.35 | 21.56 | 27.38 | 96.28 | 66.21 | 84.09 |

### 3.1. EMG Signal Acquired Using Textile Electrodes

EMG signals were collected using the developed textile electrodes and the conventional Ag/AgCl electrode for comparison via asynchronous method i.e., all types of electrodes were fixed at the same place but the EMG signal was acquired at different times. The textile electrodes were fastened on the skin using an elastic strap that has an adjustable length to keep the electrodes in the correct position. On the other hand, the commercial wet Ag/AgCl electrodes were attached to the skin with their self-adhesive pads. Three female and three male volunteer students from the Jimma University, Jimma technology institute, were recruited for this study. Each participant was allowed to partake in this study only after being briefed, reading, and voluntarily agreeing to the informed consent form. All the collected EMG signals were analyzed using the software EMG viewer manager. The protocol of this study was approved by Jimma university, Jimma technology Institutes of Institutional Review Board. All subjects were given written informed consent and provided permission for the publication of photographs for scientific and educational purposes.

### *3.2. Pre-Processing*

The feature extraction of EMG signals under three different conditions was analyzed using a MATLAB Simulink environment, visual inspection was performed during data capture and digital signal processing was performed afterward (Mathworks, Natick, MA, USA). Before extraction of important features, the recorded EMG signals were denoised using a high pass filter (wavelet denoising techniques) with a high-pass cut-off frequency of 20 Hz, which removed the low-frequency trends. Following which, the high-frequency artifacts were removed using a low-pass filter (wavelet denoising techniques) with a cutoff frequency 500 Hz. Figure 3 shows a sample of original and denoised signals from both biceps brachii Figure 2. and tibialis anterior. The better denoising method is selected after different types of filtering methods (Butterworth, Equripple, Infinite impulse response, Finite impulse response filter, and Wavelet multiresolution analysis filter) were evaluated. Wavelet multiresolution analysis denoising performed better with improved SNR.



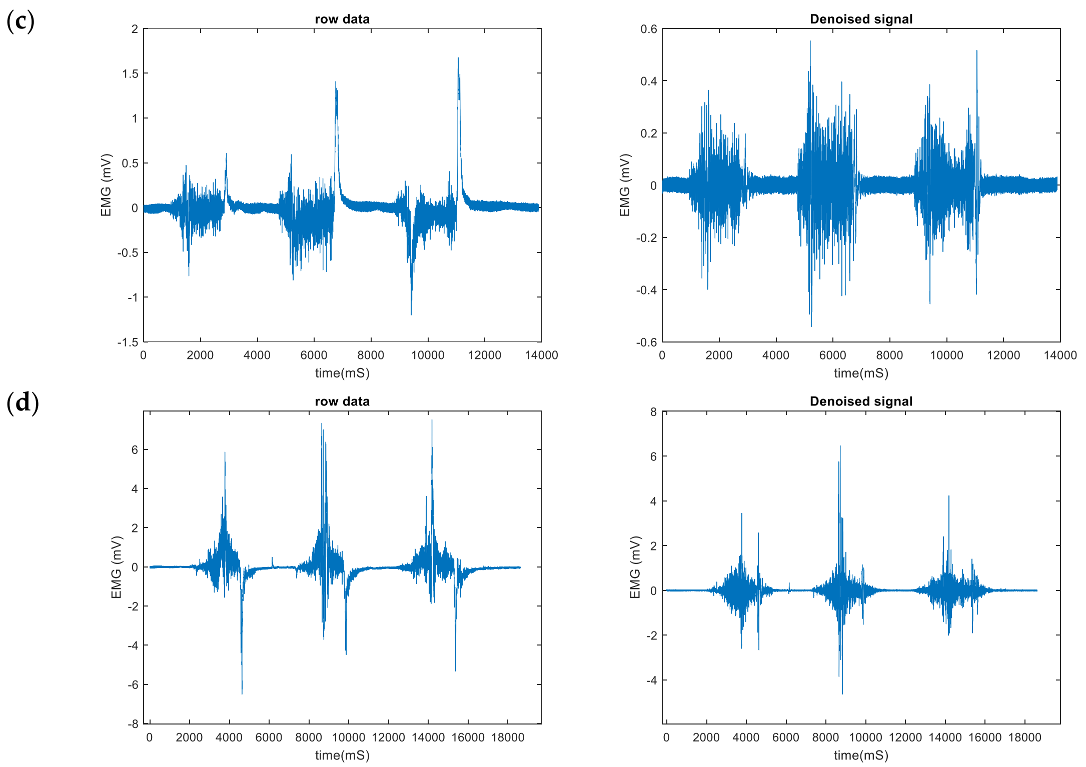
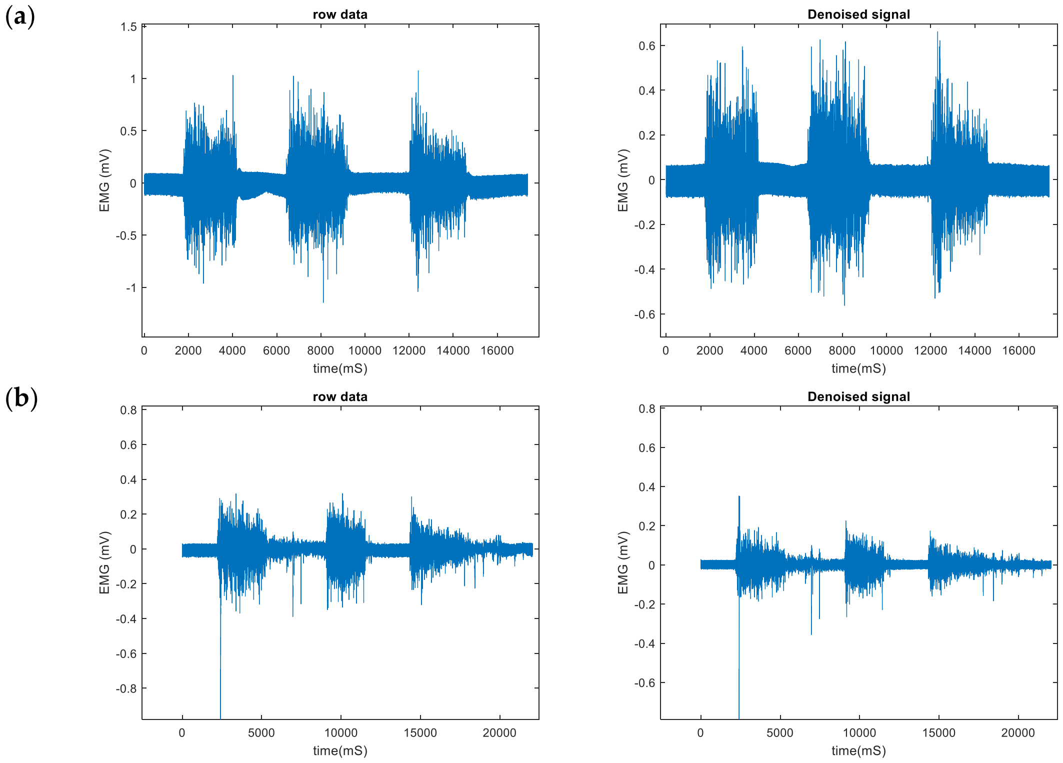


**Figure 4.** sEMG raw signal (**left**) and denoised signal (**right**) collected from biceps using (**a**) functional (Ag/AgCl) electrode, (**b**) copper platted polyester electrode (CNP) (**c**) stainless-steel electrode (SSF) and (**d**) silver platted embroidered electrode (SPEC).



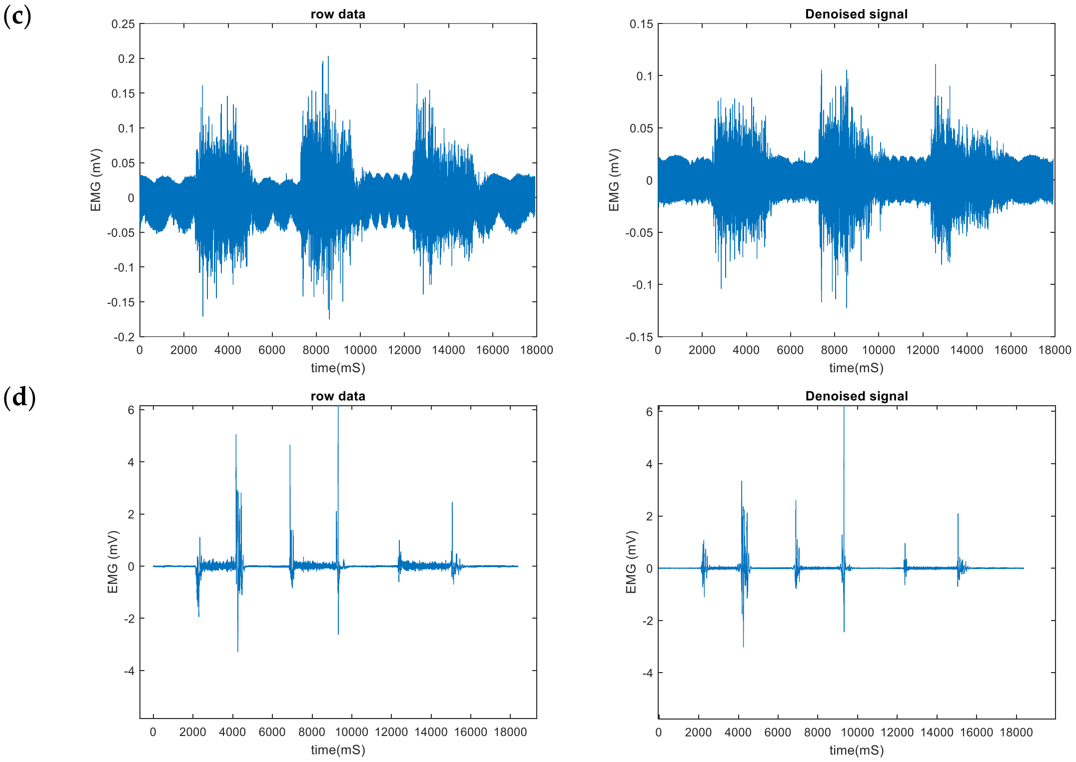


**Figure 5.** sEMG raw signal (**left**) and denoised signal (**right**) collected from tibialis using (**a**) functional (Ag/AgCl) electrode, (**b**) copper platted polyester electrode (CNP) (**c**) stainless-steel electrode (SSF) and (**d**) silver platted embroidered electrode (SPEC).

**Biplot for EMG features obtained from Tibialis:**

Figure 6 shows that the features from, AgCl and stainless steel electrodes (SFS), are clustered adjacent to each other. Whereas, the clusters of embroidery (SPEC) and polyester (CNP) were dispersed and showed significant intra-subject variations for the tibialis muscle.

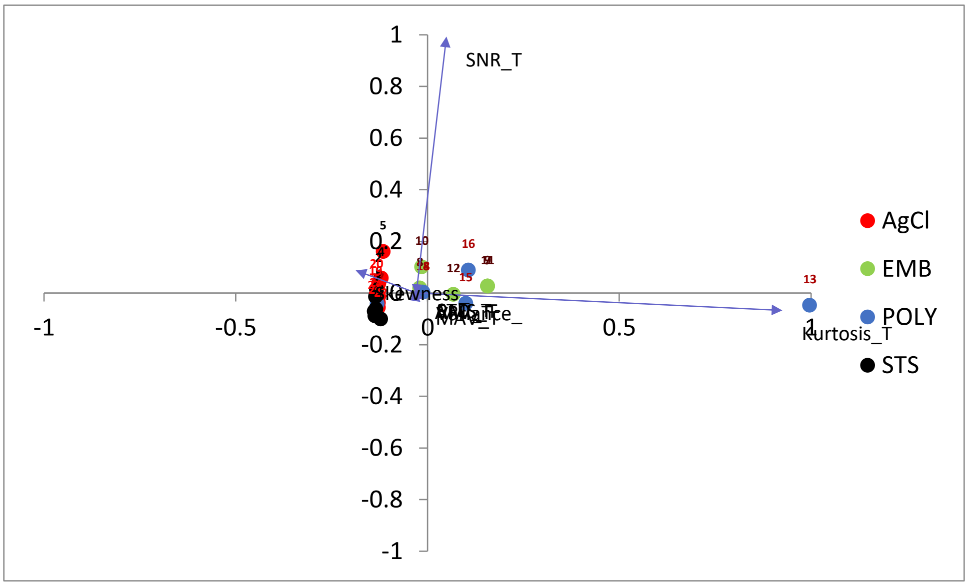


**Figure 6.** Biplot for EMG features obtained from Tibialis. The scores of each electrodes, the SPEC, CNP, SSF electrodes have been referred to as EMB (Embroidered), POLY (Polyester), STS (Stainless steel), respectively are shown in filled circles, and the loading vectors responsible for cluster separation is shown in blue arrows. The feature (loading vector) pointing towards right side, where the

clusters of POLY, EMB are spread out, i.e., 'Kurtosis', is attributed as the contributor for separation from the clusters of STS or AgCl.

**Biplot for EMG features obtained from Biceps:**

Figure 7 shows that the features from all the electrodes, AgCl, stainless steel, embroidery and polyester showed significant intra-subject variations. The variations have resulted from the 'slippage effect' observed during the experimental measurements.

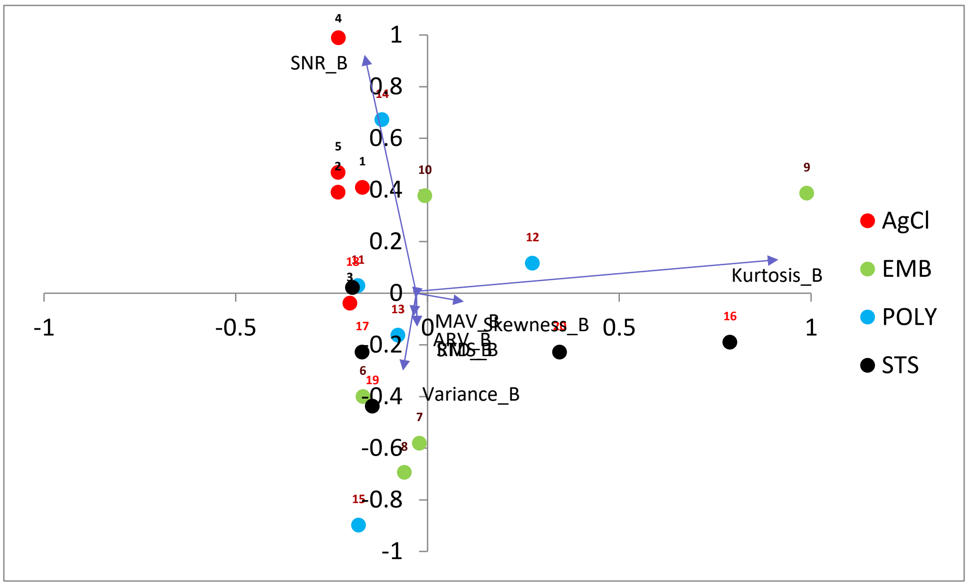


**Figure 7.** Biplot for EMG features obtained from Biceps. The SPEC, CNP, SSF electrodes have been referred to as EMB (Embroidered), POLY (Polyester), STS (Stainless steel), respectively.

## 4. Discussion

The EMG measured from six subjects show that the three types of textile electrodes recorded good signal with reasonable stability when placed on different muscles. While previous studies have evaluated single textile electrode with respect to the clinical electrode Ag/AgCl, and performed experiments on a specific muscle type and position [17], here we have used two muscle type and multiple (three) textile electrodes. Comparing different types of textile electrode among each other allows us to analyze the results statistically, removing observational bias and add statistical significance to this study.

Regarding the material type used for electrode design, it is stated that organic polymers such as polyproline are low weight, flexible and inexpensive. But they have low, and unstable electrical conductivity when compared to inorganic metals such as stainless steel, silver, and cupper-nickel composite used in this studies. This implies the need for the development of hybrid materials combining inorganic and organic components to achieve both stretchability and conductivity. In 2009 a study was conducted on eight different textile electrodes and five textile conductors [4], shows that the impedance of the electrode depends on the proportion of the conductive material present within the textile. Different yarn materials and manufacturing processes also influence the contact impedance property [3]. As there exist different manufacturing techniques; namely embroidery, knitting, and weaving; in this study, we used knitting for manufacturing both the textile electrodes (except CNP which was woven) and the substrate fabric material, on to which the textile electrode was cut and sewn (stiching).

Previously, Ozberk Ozturk et al. studied the performance evaluation of sEMG with wearable graphene textiles electrode against clinical Ag/AgCl electrodes [18]. In this study, we used the same parameters such as electrode configuration, size, shape and interelectrode distance for all the EMG signal acquisition using different textile electrodes. It has been reported in earlier studies that the EMG signal is highly affected by the size of the electrode. According to SENIAM, electrode size is defined as the size of the conductive area of a SEMG electrode [19]. In 2016, a study on comparison of different sizes of circular electrodes concluded that the larger electrodes are better [20]. The result also showed that an increase in the area will decrease resistance. Similarly, another study compared three rectangular electrodes of different sizes [21]and concluded that larger dry electrodes achieved similar performance to that of the conventional ones. In 2019, a comparative study on textile electrodes with 24 mm and 10 mm diameters [22] showed that the larger electrode exhibited low skin-to-electrode impedance. Unlike the other studies, this work revealed that large electrodes are prone to high cross-talk effects among measured signals. A more recent study carried out in 2020 on six different sizes of textile electrodes ranging from 5 mm up to 30 mm in diameter [23] showed that electrodes >20 mm exhibited lower baseline noise levels and greater SNR. However, the electrode with a 20 mm diameter exhibited better SNR value than the 30 mm one. The result clearly indicate that larger electrode (>20 mm) show lower impedance but on much larger electrodes (>25 mm) we observe more cross talks. In all these studies and similar related works, it was shown that the shape of the electrodes has less effect on the SEMG signal. While using larger electrodes, the IED will also increase in order to avoid overlapping of electrodes. According to SENIAM, inter-electrode distance is defined as the center-to-center distance between the conductive areas of two bipolar electrodes. Studies show that increasing IED will increase crosstalk. It has been suggested that unstable signal recordings due to tendon and motor endplate effects can be avoided by adjusting the inter electrode distance. Therefore, the researchers in that work prepared circular textile electrode with 20 mm diameter and used 30 mm of IED in a bipolar electrode configuration. Bipolar electrode configuration, which uses two electrodes is the most commonly used electrode configuration for sEMG, unlike monopolar and multipolar configurations [24].

Subsequent to this work, we would like to integrate the textile electrodes into a garment to asses the effect of motion artifact during movement. As reported by Alper Cömert and Jari Hyttine, a challenge for any textile electrodes is maintaining a stable skin contact during movement [25]. In our case, all our developed electrodes strapped to a conformal fit enabled recording of clean signals, even when hands were moved during load lift for biceps muscle or when running/walking on a treadmill for tibialis muscle. The stainless-steel electrode performed better than embroidered or polyester electrodes; further its performance was comparable to the reference Ag/AgCl electrode used in clinical settings. The ridges present in the SFS electrode would have assisted in maintaining a better skin contact by pressing it onto the skin and preventing its slippage from its initial position. However, the silver-coated thread used for embroidery electrode is softer when touched than the stainless steel or polyester electrode. Softness surface property of a textile electrode is much desired for its enhanced and long term wearability suitable for continuous signal measurement. Additionally, spiking of artificial sweat on the textile electrodes before signal measurement will also enhance the signal quality [26].

## 5. Conclusions

In this article, we have compared the EMG signals acquired from the biceps brachii, and tibialis anterior muscle using different types of textiles electrodes. For all the test electrodes, the conventional Ag/AgCl electrode was used as the reference standard. Morphological analysis of the time domain signal revealed that, despite smaller amplitude, the fabricated embroidered woven textile electrodes are capable of measuring the EMG signals similar to that of stainless steel and copper-nickel-plated polyester electro-conductive fabric for the same electrode size and shape. We can also conclude that the stainless-steel embroidered textile electrodes have a good measuring performance under isometric conditions, allowing the measurement of EMG signals with morphological features similar to that of the standard electrodes. This suggest that the embroidery textile electrodes may be used as a reliable alternative to stainless steel, copper-nickel-plated electro-conductive fabric, and the conventional silver-chloride. The overall noise content did not differ substantially among the electrodes and did not compromise the sensitivity of the textile electrode's signals. Further studies are in progress to evaluate the frequency content of the signals which may vary due to the different impedance characteristics of the electrodes. Further the amplitude and the phase features at different frequencies provide more insight on the dynamic activities of the muscle.

**Future work**

Comparison of EMG signals recorded from healthy and diseased/ patient individuals with the best-performed textile electrode device is in progress. Proof of concept prototypes will be developed with applications to tracking health monitoring on patients. Globally more than one million population are disabled which is approximately 15% total world population. With a large population of disabled people worldwide, many assistive technologies and techniques are in great demand to improve the quality of life for such people. World health organization has (WHO) has proposed an action plan that emphasizes the development of assistive technology [27]. Our study which has used the three textile electrodes on two different muscle types is in line with meeting the demand of the growing need for improvements in wearable health monitoring technology.

**Author Contributions:** B.B.E. designed and conducted the experiment, analyzed results, and wrote the paper; B.M. drafted the outline, analyzed experimental results; B.M. and edited the paper; J.K. helped in experimental work and result analysis and L.V.L. supervised and administered the project. All authors have read and agreed to the published version of the manuscript.

**Funding:** The research and APC were funded by the NASCERE project.

**Acknowledgments:** The authors would like to express appreciation for the support of the NASCERE project and Ghent University. The NASCERE project was sponsored with funds from the Ethiopian government. The study's design, data collection, analysis, or interpretation; the preparation of the paper; and the choice to publish the findings were all made independently of the funders.

**Conflicts of Interest:** The authors declare no conflict of interest.

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