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Embroidered archimedean spiral electrodes for contactless prosthetic control

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Abstract—With continuous advancements on active prosthetics the detection of the user's intention becomes the new technological bottleneck. While electromyography (EMG) is widely used to detect individual muscular contributions, sweat and relative sensor movements degrade the quality of the signal over time. In this paper, we bypass the problems created with the skin contact analyzing the muscular activation with Archimedean Spiral (AS) electrodes. We compare traditional EMG electrodes with AS electrodes, stacked up in textile embroidered layers to improve their functionality, and eventually adding a layer of cloth/silicon between the electrodes and the human skin to ascertain the feasibility of the method. We use $n=9$ volunteers to perform a loaded wrist extension and record signals from the extensor digitorum muscle group. We evaluate the amplitude and noise from all results and conclude that the AS electrode is capable of detecting muscular activation without touching the skin. As part of a low-cost prosthetic initiative, this EMG alternative can be potentially embedded to the prosthetic socket to improve usage and reduce adaptation problems.

I. INTRODUCTION

As active prosthetics advance in degrees of freedom and functionalities they require new ways to classify different commands from the user. The adoption of a higher number of EMG sensors to sample more muscles, or algorithms to differentiate among contraction intensities within the same muscle, could help increase the total number of output signals. To this intent the chosen EMG sensor needs to prevent undue noise from tainting the prosthetic actuation.

The most common types of surface electrodes are wet and dry electrodes. Wet electrodes, namely Ag/AgCl electrodes, are among the most used electrodes for bioelectric applications. They are disposable, light, easy and simple to use [1]. However, they require an electrolytic gel to be applied between the skin and the electrode to maintain conductivity. As the assembly dehydrates the contact with the skin is jeopardized and noise levels surge [2]. As such, EMG is generally unsuitable for long-term continuous measurement [3]. Dry electrodes do not pose such problems and can be easily applied to the skin, but they suffer from their relative movement with the skin [3] and high impedance due to the lack of the electrolytic gel [4]. Additionally, both types of

sEMG are prone to disturbances caused by the introduction of sweat in the system.

In the last ten years, new methods to reliably acquire data from muscular contractions have been investigated. Magnetomyography (MMG) has proven effective to measure the magnetic field produced by the Biot Savart law and this method has been used to register uterine muscular contractions [5]. Unfortunately, as MMG signals range from pico (10^{-12}) to femto (10^{-15}) T, these need to be shielded from earth's magnetic fields, which can reach values in the order of 10^{-6} T [6]. Radiomyography (RMG), on the other hand, aims to detect changes in the electromagnetic (EM) waves received by a sensor in proximity of a contracting muscle [7]. Some applications of RMG measure the influence that the change in muscular thickness produces on the EM wave that traverses that muscle [8], while others use a numerical computation of an AS antenna radiating into a medium with muscular properties [9]. In the latter, authors suggest that “in practice, the human body is very likely to be in the reactive and/or radiative near field of the surface mounted antenna” and their results show that AS antennas could detect a tight pencil beam radiation pattern.

In this paper, we compare conventional EMG electrodes (i.e. Ag/AgCl electrodes) with embroidered Archimedean Spiral antenna-shaped electrodes to assess the possibility of measuring contractions without contact with the skin. We place these antenna electrodes against the skin of nine volunteers and add layers of antennas to analyze the impact on the gain. Additionally, we conduct similar test with a piece of cloth and with a layer of silicon between the antenna and the skin and measure the feasibility of the sensor as part of a prosthetic socket assembly. Although the silicon layer proved too thick to result in any detectable signal, our experimental setup was capable of detecting contractions of the extensor digitorum muscle group with and without the cloth. As part of a low-cost prosthetic initiative, this EMG alternative can detect muscle contractions without being in contact with the patient's skin, and this sensor will be embedded to a prosthetic socket to improve usability and reduce adaptation problems.

II. MATERIALS AND METHODS

There are many different types of electrodes used by amputees in the myoelectric process. In this regard, the investigations focused mainly on contact and non-contact embroidered electrodes.

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Fig. 1. On the left, a conventional EMG sensor. On the right, the suggested embroidery with two archimedean spiral antennas. During experiments, we stacked up layers of these antennas with a small cloth between them to study their efficacy.

A. Electrodes

Contact electrodes are used where there is in fact little noise recording [10]. Additionally, these contact electrodes are not polarized and have a steady half-cell. However, there are large amounts of excess noise present in recordings in biological signals. The source of extra noise is as a result of the equilibrium processes localized at the surface of the skin and residual EMG signal from nearby muscles [11]. The skin was abraded in order to lower extra noise. The electrode type used determines the level of noise. In addition, amplifiers, capacitive and inductive interference and motion artifacts also reduce the signal noise ratio. One example of a conventional contact EMG electrode used in our investigation is shown in Fig. 1.

An antenna is recognised as a device with identical impedance and can be used to link the transmitter to any medium [12]. These are electromagnetic radiators, which generate an electromagnetic field between the transmitting to the receivers antenna [12]. Electromagnetic waves are converted into electrical signals that can be utilized at the initial stages. Antennas have three broad categories namely Omni-directional, directional and semi-directional [12]. These electrodes were free of noise, easy to use and affordable. As a result, these electrodes were used in the experiment to determine the correlation between EMGs and antenna, taking into consideration the AS antenna configuration. Depending on the type of application, different electrodes and electrode placing might be used.

The type of embroidered electrodes that were used in this experiment is shown in Fig 1. These electrodes were made up of the combination of sewn electrodes and cloth as the conducting layer. The polar equation used to define an Archimedean spiral is:

$$r = a\theta$$

where r is the radius, a the spiral constant, and θ the winding angle. The spiral constant a enforces a constant distance between the windings. We adopted the value of $a = 1.54\text{mm}$ and our proposed antenna reached an outside diameter of approximately 20mm with six spiral turns, in a design that resembled the simulations from [9]. The adopted thread is commercially available as a Sparkfun conductive thread

Fig. 2. Demonstration of movements performed by volunteers with a 2 kg dumbbell. In the initial position (a) the extensor digitorum muscle group is relaxed, and the contraction (b) of this group can be recorded with sensors placed on the forearm.

DEV-11791 with $3.28\Omega\text{m}^{-1}$. We adopted a Pfaff Creative 3.0 sewing machine to sew the pattern onto the fabric, similar to other previously embroidered electrodes [13].

B. Methods

The experiment was carried out to 1. find a correlation between the EMG electrodes and antennas. In addition, the aim was 2. to obtain the number of layers that the embroidered electrodes will need for best readings in an antenna configuration. The experiments took place at the University of KwaZulu-Natal and ethical approval was granted for these experiments. Volunteers were asked to lift a 2 kg weight so to minimise errors obtained in carrying out the experiment.

The following steps were carefully followed sequentially during the experiment so as to get correct readings from the digital oscilloscope:

- We placed electrodes on one of the forearm flexion-extension muscles of each volunteer.
- The volunteer kept his forearm parallel to the floor and facing downwards, as shown in Fig. 2.
- The volunteer was then given 2 kg to lift, and the change in signal peak and amplitude were registered.

This experiment was repeated with nine different volunteers, where four of the volunteers were females and five were males, and they were ranging between 20 and 33 years. Only one weight was used for each trial of the experiment, and the volunteers were required to repeat each trial twice to find a noise-free reading. The experiment comprises

Fig. 3. On the left a sheet of silicon to interface the human arm with the prosthetic limb, while on the right an example of a socket used for prosthetic purposes.

trials with the conventional EMG, with 1-5 layers of the embroidered AS antenna, with 1-5 layers of the embroidered AS antenna with a piece of cloth between the antenna and the skin, and with 1-5 layers of the embroidered AS antenna with a 2.5 mm thick sheet of silicon between the antenna and skin. This sheet of silicon is commonly used when adjusting prosthetic limbs to the human body, and it is shown in Fig. 3

C. Data Analysis

The comparison of the conventional EMG electrodes with the embroidery sensor was based on the observed amplitude of voltage (Vpp) and positive voltage (V+) from sensors during muscular contraction. An example of our experimental setting and the influence of the muscular contraction on the sensors can be seen in Fig. 4.

We adopted the Chauvenet's criterion in this experiment to find the probability band of the results obtained from each volunteer. The mean and standard deviation of the observed V+ and Vp-p from each volunteer was calculated. We used a normal distribution function to determine the probability that a given data point belongs to the result's Gaussian distribution, and any datum located outside our confidence interval was treated as an outlier and deemed spurious. The criterion was used on the embroidery AS electrodes in order to determine the best electrode layer.

III. RESULTS

The results of our experiments are identified as EMG for the conventional EMG cases and EAS for the embroidered AS cases. The addition of cloth (-C) or silicon (-S) translates as a suffix to the acronym.

A. Conventional Electrodes

Preliminary tests with EMG electrodes in different diameters have shown that smaller EMG electrodes outperform bigger ones, and the bigger diameter of the 36mm electrode might mean that it is also more prone to disturbances

Fig. 4. Initial (a) and final (b) position of experiment with the AS antenna and a small piece of cloth between the skin and the antenna. The system detects the EM influence of the muscle and the signal pattern is different from the pattern usually observed in EMG experiments.

(Fig. 5). As EMG sensors record data from the surface of the skin, the 22mm proved to have a higher amplitude and maximum positive voltage. Nonetheless, applications of these electrodes within prosthetics are very precise for the first few hours of experiment, independently of size.

B. Embroidery Electrodes

The experiment was done to determine the best combination of the different layered embroidered AS electrodes. Direct contact with the skin was adopted to determine the effect of the embroidered electrodes on the forearm muscles, and we also tried adding layers of antennas to see how it influences the gain. Fig. 6 (a) shows the results that were obtained in a setting without cloth for all the nine volunteers. The results from Fig. 6 (b), however, are based on the results

Fig. 5. Preliminary tests with 22mm and 36mm EMG sensors. The results for the smaller 22 mm electrode outperformed the bigger 36 mm electrode, and this may probably be a consequence of the interference from nearby muscles during contraction.

Fig. 6. Results from nine volunteers with five different layer settings for the EAS sensors (a), and the same V+ results for the EAS sensors without the volunteers 4 and 7 (b). The Chauvenet's criterion classified those two volunteers as outliers to the mean of the normal distribution. Dashed and dotted lines are used to facilitate visualization.

from (a) after the Chauvenet's criterion was applied. Dubious results from volunteers 4 and 7 were eliminated. The graph is showing a slight change in voltage values, which concludes that the number of layers has an effect on the output signal from embroidered electrodes.

Fig. 7. The results from EAS-C showed an improved signal quality when two layers were adopted, but a higher degree of uncertainty with the addition of more layers. As the addition of layers is also responsible for higher production costs solutions with less layers were preferred. Dashed and dotted lines are used to facilitate visualization.

C. Embroidery, Cloth and Silicon

The next step of our experimental setting consisted of adding a cloth between the electrodes and the volunteer's skin (also know as EAS-C), and this indirect contact was found to change the way the sensors behaved with the volunteers. The post-Chauvenet results of EAS-C for different layers of electrodes is shown in figure 7. The voltage values increase between the layer 1 and layer 2 on all the volunteers. In addition, there is a slight change in voltage between layer 4 and layer 5. The positive voltage values increase as the number of layers of embroidered electrodes increases, but the uncertainty of the results is too high, with little predictability of results between volunteers. In this sense, we found that the EAS-C gives the best results on average with the second layer, due to the high average signal, higher predictability and relatively lower cost. The five layers indicate a small improvement compared to two layers, yet the average improvement is 0.183 V. The cost factor has to be considered to justify more layers (higher cost) are needed for the small average value obtained with the increase in the voltage.

Furthermore, one volunteer tested the effect of placing the developed EAS electrode on top of a silicon layer (EAS-S), as seen in Fig. 8. The experiment with this volunteer showed that the signal decreases significantly when traversing a thicker medium, such as silicon, and our future works will aim to tune the design parameters of our sensor to improve the quality of this signal.

D. EAS vs EAS-C

The average V+ and Vp-p were critical parameters in determining the effect of thickness on the embroidered electrodes. EAS electrode averages are shown in figure 9. The parameters were observed and analysed as follows:

- EAS Vp-p (orange): The values decreases from approximately 1.0 V to 0.4 V between layer 1 and layer 3, and starts to increase approximately from 0.40 V to 1.50 V

Fig. 8. Our experiment with one volunteer with the EAS-S electrodes showed that the two-layered antenna produced the best result one more time, although the amplitude of the measured signal was significantly lower than the majority of cases observed with EAS-C cases.

Fig. 9. Comparison of means of EAS and EAS-C sensors, based on results from Fig. 5, 6 and 7. Higher values of V+ in relation to their respective Vp-p indicate that the muscular contraction are characterized as signal spikes, which facilitates their recognition in the midst of noise levels.

between layer 3 and layer 5. The highest average voltage corresponds to a five-layered sensor.

- EAS V+ (purple): It decreases approximately from 0.6V to 0.24 V between layer 1 and layer 3, and starts to increase approximately from 0.24 V to approximately 0.90V. The highest average voltage once again corresponds to a five-layered system.

In addition, the average voltage value (V+) for the EAS-C (dotted blue) changes from 0.7 V to 1.3 V, as seen in Fig. 9. Furthermore, the EAS-C Vp-p (dashed gray) ranges approximately from 1.1 V to 1.9 V. Both V+ and Vp-p values obtained from these experiments point to the possibility of inferring muscular contraction with or without cloth between our proposed AS electrodes. In a comparison with conventional EMG sensors, the ratio between the EMG's V+ and the EMG's Vp-p (0.088/0.172) is 51.2% and EAS demonstrated a relatively higher positive amplitude for both EAS (56.1%) and EAS-C (64.7%) cases.

IV. DISCUSSION AND CONCLUSION

This research aimed to assess the efficacy of an antenna-based sensor to detect muscle contractions. We compare different antenna configurations by adding/removing contact with the skin, adding layers of embroidered antennas and adding a 2.5 mm layer of silicon between the skin and the electrode. The results from this investigation shows that embroidery electrodes work best with two layers of electrodes stacked up. Our Figures 5, 6, 7 and 8 show a comparison of peak voltages from volunteers lifting 2 kg with the aforementioned sensor configurations, and with a convention EMG sensor as a comparison.

All results with AS antennas show that the signal varies with an increase in the number of layers, but EAS and EAS-C showed different performances with such increase. While EAS demonstrated a weak performance with 3 and 4 layered assemblies, the results for EAS-C improved proportionally to the increase in number of layers. However, the scattered results from Fig. 7 point to a low predictability of signals among volunteers, and the results with less layers presented a higher agreement among volunteers. Our experiment with EAS-S (Fig. 8) also pointed to less layers as a better solution to measure signals through a thick sheet of silicon. While the work of [9] shows that these antennas are capable of detecting signals at a 10 mm depth, in our work we proved that a real world application of similar antennas is possible at 2~3 mm away from the human skin.

A comparison between EAS and EMG sensors (Figures 5 and 9) shows that voltage spikes produced by the muscular contraction are proportionally higher for both EAS settings, while the V+ from EMG sensors was closer to half of the overall amplitude of the output signal (Vp-p).

When developing sensors total cost might be an important design criteria, and this cost can be reduced with a lower number of layers. The combination of a lower production cost, the highest V+ achieved with EAS-S experiments and a higher predictability of user contraction during EAS and EAS-C experiment makes the two-layered EAS sensor the sensor of choice for our prosthetic system.

Non-contact electrodes have many advantages over contact electrodes: The lack of contact with the skin reduces the influence of sweat in the measurement, increases the possibility of embedding these sensors to prosthetic sockets and can be easily adapted to be used on top of clothes. Additionally, non-contact measurement is also suitable for long-term measurements since conductive gel desiccation can result in loss of contact between the electrodes. Less noisy and smoother signals were observed for non-contact electrodes than contact electrodes. The features implementation of non-contact electrodes for amputees have been obtained, because of their sophisticated, comfortable and non-obtrusive EMG measurements and it can be concluded that there is value in using them.

In our future works we will test similar electrodes to control our prosthetic hand Touch Hand III and explore ways to discretise contraction intensities with this electrode.

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