Embroidered Electromyography: A Systematic Design Guide

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Ali Shafti, Roger B. Ribas Manero, Amanda M. Borg, Kaspar Althoefer, *Member, IEEE*, and Matthew J. Howard, *Member, IEEE*

***Abstract*— Muscle activity monitoring or Electromyography (EMG) is a useful tool. However, EMG is typically invasive, expensive and difficult to use for untrained users. A possible solution is textile-based surface EMG (sEMG) integrated into clothing as a wearable device. This is, however, challenging due to (i) uncertainties in the electrical properties of conductive threads used for electrodes, (ii) imprecise fabrication technologies (e.g., embroidery, sewing), and (iii) lack of standardization in design variable selection. This paper, for the first time, provides a design guide for such sensors by performing a thorough examination of the effect of design variables on sEMG signal quality. Results show that imprecisions in digital embroidery lead to a trade-off between low electrode impedance and high manufacturing consistency. An optimum set of variables for this trade-off is identified and tested with sEMG during a variable force isometric grip exercise with n=12 participants, compared with conventional gel-based electrodes. Results show that thread-based electrodes provide a similar level of sensitivity to force variation as gel-based electrodes with about 90% correlation to expected linear behavior. As proof of concept, jogging leggings with integrated embroidered sEMG are made and successfully tested for detection of muscle fatigue while running on different surfaces.**

***Index Terms*— Electromyography, Wearable Sensors, Biomedical Monitoring.**

I. Introduction

THERE are commercial wearable devices available today, measuring the number of steps, heart rate and sleep activity over extended periods. However, wearable continuous monitoring of electro-physiological signals, e.g., brain or muscle activity, is still undergoing research. Non-invasive recording of muscle activity requires the use of surface electromyography (sEMG) whereby pairs of electrodes are applied to each muscle of interest within a muscle group, and a single ground electrode placed on an unrelated, preferably

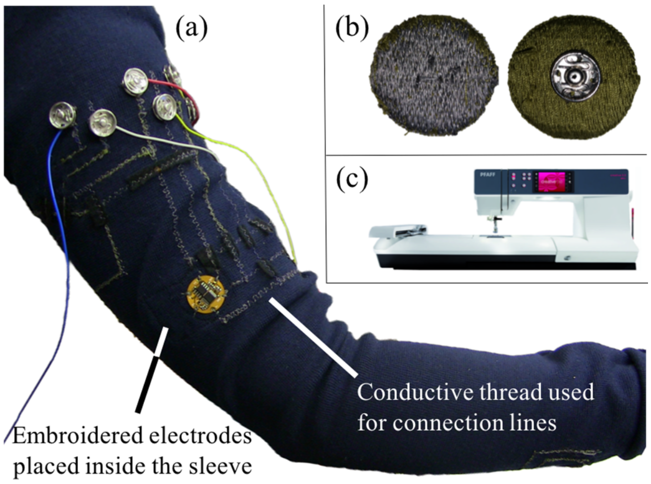

Fig. 1. Embroidered electrodes and their use in wearable sEMG. (a) Sleeve equipped with textile based sEMG, (b) Embroidered electrodes used for the sleeve, (c) Pfaff Creative 3.0 digital embroidery machine used to manufacture the electrodes – machine photo courtesy of PFAFF

muscle-free, part of the skin. Proper placement of electrodes is challenging for an untrained user, as the location and spacing between electrodes affects the resulting sEMG signal. However, if the sEMG sensors are integrated into a wearable platform, such as clothing and textile (refer to Fig. 1) the user will only have to wear the textile as they normally would and the sensors will already be located in the correct positions. The need for reduction of line noise and motion artefacts define the challenges in designing appropriate electrodes, electronics and signal processing for such systems.

This paper reports on a step-by-step approach to realizing textile-based sEMG electrodes. Different design variables concerning the fabrication of electrodes through embroidery are examined, and the manner in which they affect the electrical characteristics of the electrodes assessed. The behavior of the embroidered electrodes during actual sEMG acquisition is evaluated, in comparison with typical gel-based electrodes used in medical applications. In line with expectations, it is seen that measuring sEMG with embroidered electrodes encounters higher noise levels compared to gel electrodes, however, the variation in force is still distinguishable suggesting the feasibility of using embroidered electrodes in applications such as effort and gait analysis while providing a higher level of comfort and ease of use. This paper therefore presents, for the first time, a study to characterize the electrical properties and manufacturability of

This work was supported by the UK Crafts Council as part of the Parallel Practices project and by the Seventh Framework Program of the European Commission under grant agreement 287728 in the framework of EU project STIFF-FLOP and by the Horizon 2020 Research and Innovation Program under grant agreement 637095 in the framework of EU project FourByThree.

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conductive textile based sEMG electrodes and shows their use in jogging leggings which are worn and tested outdoors as a proof of concept. A preliminary version of this work has been reported in [1].

II. BACKGROUND

A. Motivation

There are a number of benefits to be gained in developing a wearable, textile-embedded sEMG acquisition system. These include (i) continuous, remote monitoring of muscle activity, (ii) ease of use due to lack of concerns on proper placement of electrodes, (iii) comfort in wearing and unobtrusiveness. If sEMG sensors were integrated into clothing, the resulting garment would allow for continuous monitoring of muscle activity without obtrusion to the user, or difficulties in electrode placement. If a patient suffering from long term chronic and degenerative illnesses is equipped with a set of wearable sensors, their activity, movement and well-being can be monitored continuously and remotely [2]. Surface EMG can be used on the legs to monitor leg muscle activity and fatigue during exercise – this is done as proof of concept in this paper as well as used in ventilatory threshold detection during incremental running in [3]. It can also be used to monitor ergonomics through the status of the muscles in people with jobs involving continuous stances to warn of potentially unhealthy postures [4], or as an assessment tool to examine comfort in jobs involving precise tasks with special or heavy tools, long hours and unusual postures leading to a difficult lifestyle, e.g., surgeons [5]. This is better achievable through a wearable platform.

Moving into a textile substrate, however, presents a number of engineering challenges. If the system is to be properly integrated, compromises must be made on some of the shielding and noise cancellation techniques common in modern electronics. The skin-electrode interface is considered the most critical block in sEMG systems, as mismatches between the two areas of skin-electrode contact is the largest source of noise [6]. This is especially so in textile-based sEMG due to the uncertainties in fabrication of electrodes realized by embroidering conductive textile elements into fabric. Embroidery is especially highlighted here as it is added to fabric that is already made. Weaving and knitting allow for the manufacture of fabric as a whole and have been used by researchers in exploring the possibility of realizing fabric antennae or integration of electronics into fabric [7] [8]. However, this paper looks at the characteristics and behavior of electrodes made of conductive textile and can thus be applied to all the above mentioned techniques.

The use of digital embroidery is adopted in an attempt to better control the design variables and reach repeatability which is not an option with handcraft. However, even with digital embroidery tools there are issues with imprecisions and errors in manufacturing that can lead to unexpected results and lack of consistency. This paper presents the first design guide for conductive thread-based skin-surface electrodes with a thorough characterization of their electrical properties and

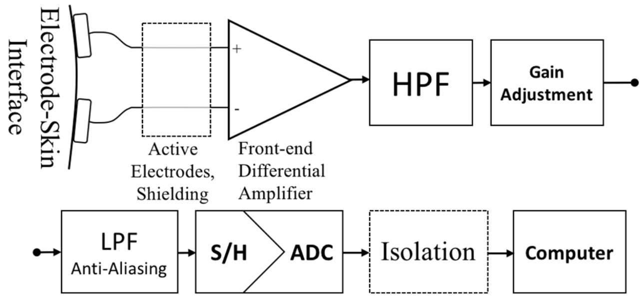

Fig. 2. Simplified block diagram of a typical sEMG acquisition system.

suitability for measuring sEMG.

B. Theoretical Background

To better understand the challenges involved with embroidered sEMG systems, a better understanding of underlying theory is required. There are issues in sEMG acquisition that are further emphasized on a wearable platform. The skin's conductivity and impedance vary from person to person and even for the same person based on the area of skin, how moist it is and how hairy it is [6]. The impedance might even evolve in the long term based on the person's lifestyle. Apart from this, using the sEMG method of picking up signals from the skin, the measured signal contains components not only from the particular muscle of interest, but also from surrounding muscles and organs, movement artefacts that cause the electrode-skin interface characteristics to change, and environmental noise such as the 50/60Hz power line noise. There are, therefore, many sources of noise and inaccuracies to be considered when designing the sEMG system.

Fig. 2 shows a typical sEMG acquisition system's block diagram which comprises of differential amplification followed by filtering. In differential amplification, two input pins are used (positive and negative). The output of the amplifier will be the difference of the signal picked up at these two pins multiplied by a certain gain defined by the circuit topology. This can be described as:

$$V_{out} = A_d\,(V_{in+} - V_{in-}) \tag{1}$$

where Ad is the differential gain of the amplifier and Vin+, Vin- and Vout are the positive input, negative input and the output pins respectively. Therefore, whatever difference exists between the two signals at the two input pins will be amplified and anything they have in common will be rejected due to the subtraction in (1). This is especially useful in the case of sEMG where the two electrodes picking up the signal have different sources of noise in common (power line, movement artefacts and cross-talk from other organs) that need to be cancelled out. However, in the case of sEMG, a good common-mode rejection is not enough to mitigate all noise. As it only ensures that the common-mode signal at the output is negligible compared to the amplified differential signal. But this is based on an assumption that the noise is present as a common-mode signal at the amplifier inputs. In the case of sEMG, this means that the connection from the two amplifier input pins all the way to the signals under the skin need to be matched in terms of impedance.

A mismatch between these two paths will lead to the noise being scaled differently and picked up with different levels at the amplifier inputs and thus not rejected; as it will look like a differential signal to the amplifier. Matching these signal paths however is not an easy task as skin impedance is unpredictable and any movement artefacts can affect it. Even the most stable electrode connections to the skin will move as the muscle is contracted and moves under the skin - more so the case when using conductive thread based electrodes integrated into clothing. This is why the electrode-skin interface is the most sensitive and important part of the sEMG system. While skin preparation, such as shaving and use of gel or alcohol is recommended [9] [10], it is not possible in a textile-based wearable system, adding further difficulty to matching the paths.

For the application of the present research, a highly accurate and noise-free signal is not the main objective. The goal is rather to acquire sEMG with enough noise cancellation to make the changes in signal due to different levels of force and effort distinguishable. As shown by this brief theoretical overview of the sEMG acquisition system, the electrode-skin interface is the most important part of the system, and thus designing it will require a thorough analysis of different variables involved and how they affect the overall interface behavior. Matching the impedance of the electrode-skin interface is particularly challenging on a textile substrate due to (i) mismatches between the two areas of skin-electrode contact, (ii) the electrodes are dry, without special skin preparation, (iii) issues such as the flexibility and stretching of the fabric may affect the resistance of the thread. The conductive thread used for the electrodes has a resistance per length specification considerably higher than conventional copper wires (91.8Ω/m for the thread compared to 0.0098Ω/m for copper wires based on their relevant datasheets). Thus, the length of the thread that is used in connections plays a larger role in defining the skin-electrode interface impedance, relative to standard electrodes. These issues have not been systematically examined in the literature.

C. *Related Work*

Many efforts have been made on wearable monitoring of muscle activity. In [11] a miniaturized sEMG system is presented and used for wearable purposes. However the device itself is not wearable nor can it be integrated into wearable devices or clothing. In [12], a wearable device for facial expression detection is presented. The wearable interface is placed around the back of the head touching the skin on the cheeks. The paper reports good results on detection of different expressions, however, is not unobtrusive enough to be worn continuously in the long term. A more focused effort on creating wearable sEMG systems is presented in [13]. However, the boards are still bulky and they are not integrated directly into clothing or a small wearable device. In [14], a multisensory system including a sEMG sensor is created in the form of a wristband, similar to a watch. This work presents a wearable device that can be used in long term applications. It is, however, different to the aim of this paper, as it is not for applications involving integration with fabric and clothing and it is also limited to the specific use on the wrist. The system developed in [15] is perhaps the one with aims most similar to that reported in this paper. The sEMG sensor is integrated into the fabric as a small circuit, using active electrodes. This work also relies on conductive thread for their circuit connections. However, they rely on capacitive detection of signals as described in [16]. While the contactless approach is useful for this sort of application, as it doesn't rely on contact with the skin, it raises some concerns into the integrity of the signals picked up, and introduces new variables into the design approach. In addition, as all sEMG acquisition and analysis methods are based on contact between the electrode and the skin, it is not clear if the contactless approach can be used with the previously validated analysis techniques and whether results obtained from these sensors will be as reliable. The present study differs by virtue of endeavoring to detect sEMG signals using direct contact with conductive thread based electrodes. This is beneficial as, if achieved, it limits the sources of noise and inaccuracies to those already identified for typical sEMG applications and does not introduce new challenges as is the case with the contactless approach. In [17], the authors compare the performance of textile electrodes made of fabric and integrated into shorts with that of conventional electrodes. The study looks at the specific case of conductive fabric electrodes made of silver fibers. The study shows the feasibility of using these fabric electrodes for EMG acquisition. It does not however consider the design of the electrodes and how they might affect the quality of the acquisition. Being fabric, the electrodes are rather large (39cm2 in the small version of the shorts) which is not in line with latest research findings recommendations. Embroidered electrodes explored in this work allow a more precise design and smaller sizes (12.5cm2 in the final design) allowing for more accurate acquisition.

To the best of our knowledge, this paper is the first study into the different design variables involved in the design of embroidered sEMG electrodes and how they affect the quality of the acquired signal.

III. METHODS AND MATERIALS

The electrodes tested in this paper are made of stainless steel conductive thread, sewn into fabric with regular thread to hold it in place. A Pfaff Creative 3.0 programmable sewing machine was used to make the electrodes. The steel thread is wound onto a bobbin and placed in the bobbin case, and the regular thread is kept on the spool and placed in the spool case. Before the fabrication process starts, the fabric is placed in an embroidery hoop. Hooping the fabric along with a stabilizer keeps an adequate tension required by the embroidery process. During this process, the regular thread is passed through the fabric from one side as it pulls the conductive thread towards the other side of the fabric following the specified design.

The electrodes were designed using the 6D Embroidery

System software provided by the sewing machine manufacturer. The tension setting, which defines how tight the thread is sewn into fabric, is set at the maximum possible without breaking the conductive thread (T=5). Higher values might not break the thread, but would lead to it going through the fabric in error. Lower tension values were not used to avoid looseness in the conductive thread which would lead to inconsistencies and potential misconnections. A stable contact between different lines of conductive thread is needed for them to be paralleled and reduce overall resistance. The same conductive thread is also used to sew 13mm diameter studs (snap fasteners) to the center of each electrode for easy connection to the rest of the circuit (see Fig. 4).

For sEMG acquisition with these electrodes, an acquisition circuit is necessary. Fig. 3 shows a diagram of the acquisition circuit used for this study. Signals are picked up from the skin using the electrodes. An active electrode system is used where a unity gain buffer is placed on top of the electrodes to assist with impedance matching between the two signal paths. Wire connections take the output of the unity gain buffers to the inputs of an instrumentation amplifier (Analog Devices AD620B). A capacitor is added in series to the gain programming resistor of the instrumentation amplifier (IA) to enforce a frequency varying gain, high pass filtering the sEMG signal as it is amplified. The IA gain is set at 28.5V/V. The output of the IA is followed by a 1st order active HPF at 20Hz to mitigate any remaining low frequency noise such as motion artefacts as well as any potential DC offset from the IA. The active HPF also applies a passband gain of 19V/V. The HPF is followed by a 1st order active LPF at 410Hz as an anti-aliasing filter. The output of this filter is ready to be sampled and converted to digital data. Thus the overall circuit gain is 541.5V/V. In order to ensure that the new sensors followed validated and established standards in sEMG acquisition, the recommendations of SENIAM were followed in selecting the above mentioned parameters. SENIAM or "Surface Electromyography for the Non-Invasive Assessment of Muscles" was a European Union project focused on setting standards for sEMG sensor design, placement and data analysis in order to unify and concentrate sEMG research methods across Europe [18]. Publications originating from the company Delsys which provides commercial EMG solutions were also used to consider factors such as frequency ranges, inter-electrode spacing, electrode placement, reference selection and electrode design during the design and application of the sensor, and the overall acquisition system [19-22].

IV. Experiments

The experiments reported in this section consist of electrical property tests on the electrodes with different sets of design variables to identify how each variable affects the electrical characteristics of the resulting electrode as well as to settle on the best design to move forward. This is followed by sEMG acquisition tests to examine the identified design in a real application.

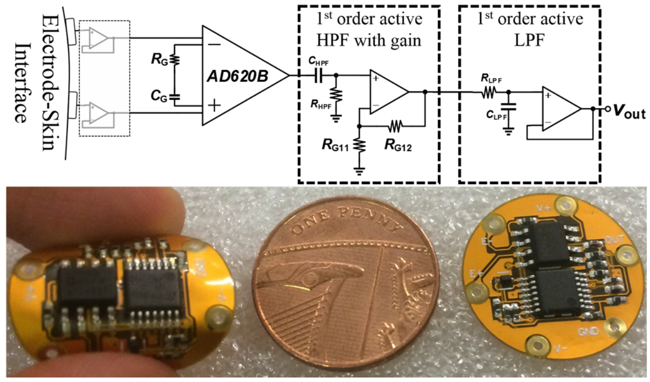

Fig. 3. sEMG acquisition circuit diagram (top) and the PCB designed for the proof of concept on the right (bottom).

A. *Electrical Property Test*

The aim of this experiment, is to identify and characterize the effect of different design parameters on the electrical (i.e., impedance) properties of the embroidered electrodes.

Surface electrodes in general are characterized by physical dimension, shape, technology and constituent materials [18] [23]. When using the sewing machine, aside from the shape and dimension of the electrodes, alternation between different fill densities (i.e., distance between two consecutive thread lines inside the pattern) and different number of iterations (i.e., how many times is the same pattern sewn) is possible.

Preliminary experiments showed similar behavior between different electrode shapes (circle, square, and rectangle), and so this factor was discarded from the study (in the following, results are reported for circular shaped electrodes). The three main variables considered are thus (i) the area, (ii) the number of iterations and (iii) the density of the fill in pattern. Fig. 4 shows these design variables and their expected effect on the overall electrode resistance.

Based on the resistance equation:

$$R = \rho\ (L/A) \qquad (2)$$

Where R is resistance in Ohms, ρ is resistivity, L is the length and A is the cross sectional area of the wire, it is expected that shorter lengths of thread and parallel connections between threads (due to paralleling of resistances) will result in lower overall resistance and thus lower impedance for the interface. This translates into smaller electrode area, higher number of iterations and higher density

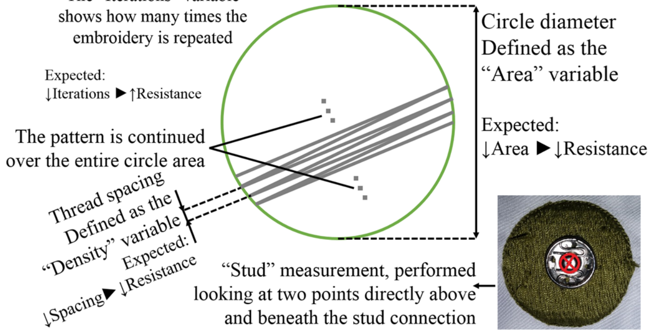

Fig. 4. The inspected design variables shown for clearer presentation. The figure represents a circular electrode during embroidery.

TABLE I

MEAN AND P-VALUES FOR RESISTANCE OF DIFFERENT ELECTRODE DESIGNS

#	Electrode	Mean	SD	P-Value	
1	(2,2,3)	0.30	0.05	-	**Baseline**
2	(**1**,2,3)	0.27	0.05	$6.54x10^{-2}$	**Area Variation**
3	(**3**,2,3)	0.22	0.04	$1.01x10^{-5}$	
4	(2,**1**,3)	0.33	0.04	$5.48x10^{-2}$	**Iteration Variation**
5	(2,**3**,3)	0.34	0.07	$2.36x10^{-2}$	
6	(2,2,**5**)	0.19	0.06	$2.71x10^{-7}$	**Density Variation**
7	(2,2,**4**)	0.35	0.08	$1.99x10^{-2}$	
8	(2,2,**2**)	0.19	0.06	$2.71x10^{-7}$	
9	(2,2,**full**)	0.19	0.03	$6.52x10^{-9}$	

as the predicted optimal electrode design.

To test this, a set of electrodes varying each of these parameters were fabricated using the Pfaff sewing machine as described in §III. With regards to (i) area, results are reported for electrodes of diameter 1cm, 2cm and 3cm, (ii) iterations, those for electrodes fabricated with 1 (no repeats), 2 and 3 iterations are reported, and (iii) density of the fill-in pattern, thread spacing of 5mm, 4mm, 3mm, 2mm and full (no spacing) are reported.

In order to see how each parameter affects the overall electrical behavior of the electrode, each parameter is varied independently, and results are compared against a baseline electrode design, representing the mean values for each variable, i.e., 2cm diameter (medium area), 2 iterations and 3mm density spacing.

To characterize the impedance properties of the electrodes, the resistance measurement as shown in Fig. 4 is measured using a digital ohmmeter. Each measurement is taken N=20 times for each electrode followed by ANOVA used to determine whether the mean resistance value varies significantly or not according to each parameter varied. The measurements are reported in Table 1 across different variations in the design variables. Electrodes are labelled using the format: (area, iterations, density), e.g., the baseline electrode -- which has a 2cm diameter, 2 iterations and a density spacing of 3mm -- is referred to as (2,2,3). The results are presented in Table 1 color coded according to which parameter is varied.

Looking at the effect of area variation, a decrease in area (line 2) does not result in a statistically significant variation in resistance, but an increase in area (line 3) results in a significant decrease in resistance. In terms of iteration variation, a decrease in iteration (line 4) makes no significant difference but an increase (line 5) results in a significant increase in resistance. During density variation, a decrease from the baseline (line 7) results in an increased resistance and an increase of density (line 8) will result in a decreased resistance. Further density increase (line 6) as well will result in decreased resistance. Further density reduction also results in decreased resistance (line 9).

The results generally follow the theoretically expected trend, but there are also some unexpected results. These anomalies can however be explained if the effects of the embroidery machine manufacturing inaccuracies are taken into account. The embroidery machine is not entirely accurate at low dimensions and spacing, leading to manufacturing errors. This results in unexpected behavior. These errors will result in the conductive thread not paralleling with itself which leads to longer series connections that could therefore result in higher overall resistance values than expected. Thus, a test on manufacturability is necessary. Based on these experiments, the designs with the best results (lowest mean impedance) are those with 2cm diameter (medium area), 2 iterations (medium number of iterations) and 2mm, 5mm or full spacing grid pattern (lines 6, 8 and 9, respectively).

B. Manufacturability Test

The anomalies in the results from §IV.A highlighted the importance of manufacturability of design variable sets and the fact that not every set can be implemented without errors. Fig. 5 shows 2 samples of the same electrode design made with the sewing machine. Errors and inconsistencies due to the small dimensions are clearly visible in the picture (Fig. 5.b). The electrode area is expected to be covered entirely with conductive thread (dark grey) whereas in some of the electrodes, the green top thread (which holds the conductive thread in place when sewn) is seen coming through the fabric and covering part of the electrode surface in error. These errors will result in inconsistencies. Apart from low impedance in the skin-electrode interface, a good impedance matching between the two paths (i. e. the two electrodes, refer to §II.B) is needed. Thus low consistency in manufacturing results will be problematic and the next issue to consider is how consistently the digital embroidery machine can manufacture a particular design variable set.

Using the best design as derived in §IV.A, here we present a manufacturability test to see if the sewing machine can create the same design repeatedly. To this end, twelve electrodes of the identified design variable sets were made and their resistance value measured N=20 times. This is done for the 2cm diameter, 2 iterations, 2mm spacing (2,2,2) and the 2cm diameter, 2 iterations, full pattern (2,2,full) electrodes. Fig. 5.a shows a whisker plot representation of the results of these measurements.

The grid pattern electrodes show a lower resistance value (Mean=0.64, Standard Deviation=0.0423, N=20) than the full

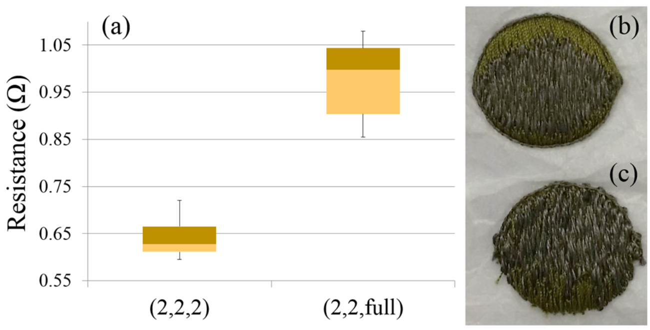

Fig. 5. Manufacturability test. (a) Resistance variation as a result of manufacturing inconsistencies for the two sets of design variables identified in §IV.A. (b) Embroidered electrode with visible errors (c) embroidered electrode with no visible errors.

pattern electrode (Mean=0.9745, Standard Deviation=0.085, N=20) along with a lower standard deviation. This is explained by the manufacturing errors made by the sewing machine. The full pattern results in crowded embroidery and threads being sewn into fabric very tight and close to each other. This leads to difficulties for the machine and makes errors unavoidable. These errors are not always the same and thus lead to inconsistencies. This is why the standard deviation of the resistance is higher in the full pattern. The grid pattern with 2mm spacing allows some breathing space for the sewing machine and relaxes the manufacturing requirements and conditions, resulting in less errors and more consistent results.

Having confirmed the effect of manufacturing errors and how they increase in lower dimensions, the anomalies in §IV.A can now be better explained. In the case of area variation, no change in resistance is expected for the measurement as it is measured between two points directly beneath each other (refer to Fig. 4). However, line 3 in Table 1 shows a significant change in resistance. Considering the small dimensions of the design and the anomalies resulting from it, it can be deduced that an increase in area will result in a more relaxed design for the embroidery machine due to the increased dimension which results in a less tight workspace for the machine. This means there are less errors in the manufactured electrode and therefore a lower resistance is observed. For iteration variation, the resistance is expected to decrease with increased number of iterations, but this is not observed in the results (Table 1, line 5). Adding another iteration of the same pattern within the small dimension of the electrodes results in a crowded design that can lead the machine into errors. These errors can explain the increased resistance. During density variation, lines 7-9 of Table 1 present the expected result of decrease in resistance with increasing density as compared to the baseline. However, a decrease in density is expected to increase the resistance, which is not the case (Table 1, line 6). As the design gets denser, manufacturing becomes more difficult and errors can be made by the embroidery machine. This can explain why a lower resistance is witnessed when the density is reduced as it results in a less complex design.

The above errors mean that in low dimensions and crowded designs, the results expected by the theoretical analysis do not match. There is a trade-off between achieving lower resistance and keeping manufacturing errors to a minimum that can also lead to inconsistencies.

Thus, based on our experiments, for electrical characteristics, the 2cm diameter (medium area), 2 iterations (medium number of iterations) and 2mm spacing grid pattern (2,2,2) provide the best results in terms of low impedance and consistency. This makes (2,2,2) the best suited design for sEMG acquisition. It must be noted that these specific results depend on the selected digital embroidery machine. The device used in this study is a high-end device, so can be considered more accurate in sewing than cheaper machines. However, the method and process presented here to verify manufacturability can be applied to any machine and is in line with the aim of this paper to present a systematic design guide for researchers in the field.

C. *sEMG Tests*

Having characterized the electrical properties of the electrodes according to the design variables (ref. §IV-A&B), in this section, their performance for sEMG acquisition is examined in comparison to conventional medical gel-based electrodes. The aim is to evaluate the use of the conductive thread-based electrodes in terms of distinguishing between different levels of force applied by the muscles.

For this, the experimental procedure is as follows. Pairs of electrodes are placed on the participant's forearm muscles, one pair on the flexor muscle group and another pair on the extensor muscles. A dynamometer (Camry 90kg Digital Hand Dynamometer) is fixed to a table as an exercise tool. The participant rests their arm on the table, and the arm is fixed in place using strapping in a position where they can grip the dynamometer with comfort. Participants are then asked to apply grip forces to the dynamometer. As the arm is fixed to the table, the exercise can be considered isometric and therefore a linear relationship between sEMG and force levels is expected [24]. The dynamometer allows the monitoring of the applied force level. The active electrode circuit used to pick up the electrode signals (Fig. 3) is housed inside a box with handles that allows it to be fixed onto the participant's arm similar to an armband. The box also houses stud connectors that can connect to both the conductive thread electrodes and the gel ones (Covidien Kendall Arbo H124SG), see Fig. 6. In this manner, stability and fair comparison between exercises is ensured. The outputs of the circuit are connected to a Bitalino microcontroller system which samples the signal at 1kHz and sends the data wirelessly through Bluetooth to a nearby computer. The signal is recorded in the computer and further analyzed using MATLAB. Fig. 6 shows the experiment setup.

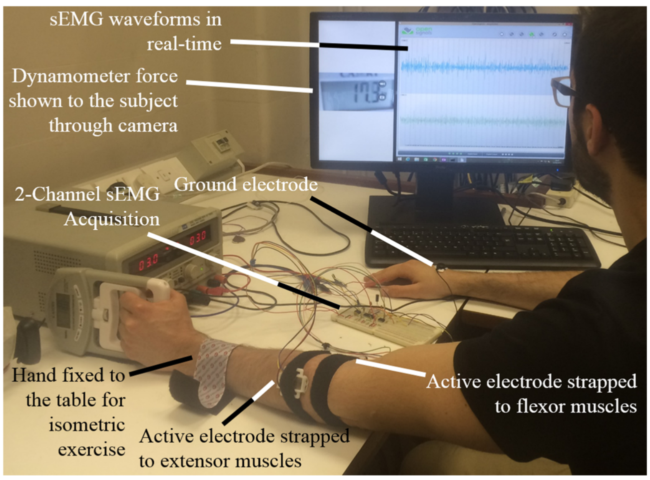

Fig. 6. Experiment setup. The active electrodes are held by hook and loop strap and are touching the participant's skin through embroidered electrodes.

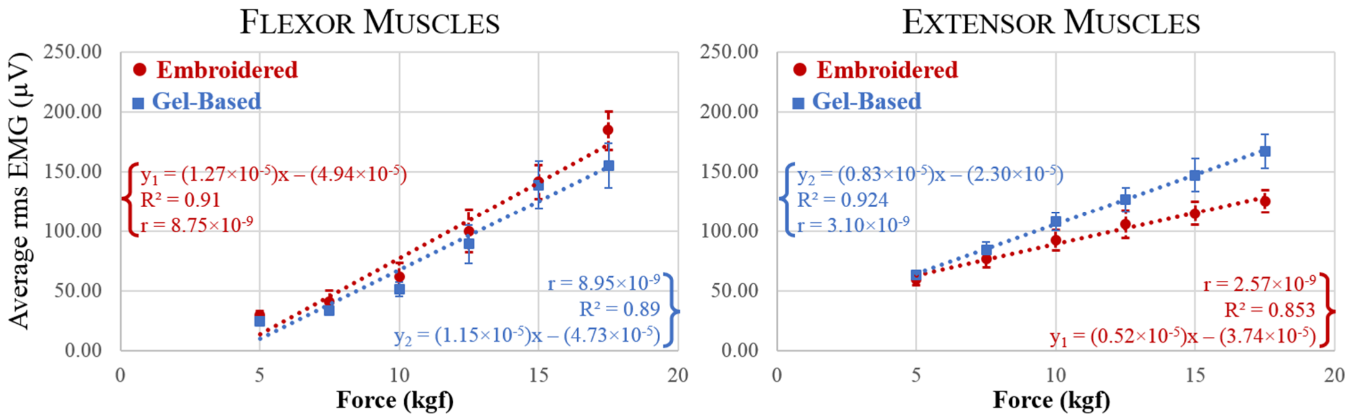

Fig. 7. EMG values versus force levels for different electrodes tested with one of the twelve participants' flexor and extensor muscle groups.

Each participant applies a set of 6 force values on the dynamometer while their muscle signals are recorded. The participants are asked to look at the force value displayed on the dynamometer and attempt to hold the 6 specific force values, each for 10 seconds, with 10 second rests in between. The force values were 5kgf, 7.5kgf, 10kgf, 12.5kgf, 15kgf and 17.5kgf[1]. This exercise is repeated 5 times with the embroidered electrodes and 5 times with the gel-based electrodes – 10 exercises overall – with 5 minute rests in between exercises to avoid fatigue. The results reported here are those obtained from n=12 participants (8 male, 4 female). Participants vary in age (23±6 years old), gender, ethnicity, dominant hand and hours of exercise per week. Ethical approval was obtained prior to the tests (reference number BDM/13/14-123).

From the raw data, basic pre-processing is applied to get a clear signal. First, the signal is divided into equally sized sections corresponding to each single force application and the mean value of each is removed (i.e., the DC set to 0). The resulting signal is then scaled and attenuated to show the actual sEMG value on the skin interface before the circuit gain is applied. This is then high pass filtered using a 4th order Butterworth filter at 20Hz. The filtered signal is rectified by obtaining its absolute value and smoothed by computing the root mean square (rms) of values in a sliding window to obtain a linear envelope. The window size used for here is 200ms. Finally, the average value for the linear envelope during the force application is calculated and used as a measure of the sEMG level for that particular force. In this manner, each single force value has a corresponding single sEMG value associated to it.

Fig. 7 shows the sEMG versus applied force data for thread (in red) and gel (in blue) electrodes, and a trend line for one participant. As can be seen, the gradient of the thread-based and gel-based trend lines are quite close to each other. The trend line for the conductive thread based electrodes has a 12.4% higher gradient when compared to the gel electrodes in the flexor group, showing more sensitivity to force variation. This is 40.1% lower in the extensor group, where gel electrodes have shown a higher sensitivity to force variation when compared to the thread electrodes. R-squared is calculated to verify the correlation between the EMG and the measured force, showing 87% correlation for the thread-based electrodes and 88% for the gel-based in the flexor group. Sum of squared residuals (r) is also given in Fig. 7. These results show that the conductive-thread based electrodes have been effective in providing similar results to those of gel based electrodes.

Table 2 summarizes the results of the experiment considering all participants. The results of participant 7 were discarded, as the recorded EMG data from their extensor muscle group was corrupted during three of the trials; these results are therefore for 11 subjects. The gradient of the trend lines is similar; 10% higher for the thread electrodes in the flexor group compared to the gel electrodes. R-squared results show 44.2% correlation for the conductive thread-based electrodes and 46.1% for the gel-based in the flexor muscle group. Similar results can be seen for the extensor group however with lower correlation values and in this case a 23% higher gradient value for the gel-based trend line compared to the thread based. Overall, results show that the different force levels can still be distinguished using the conductive thread-based electrodes prepared as part of this study.

Similar behavior was expected for the flexor and extensor muscle groups. However, in the case of the extensor muscles, the thread electrodes are showing worse performance when compared to gel electrodes. This might be due to the fact that the side of the forearm where the extensor muscles are located is typically hairier, resulting in more noise for the dry thread electrodes when compared to gel electrodes (that adhere to the skin even in the presence of hair).

These experiments confirm the ability of embroidered electrodes to acquire sEMG signals under isometric conditions in a laboratory setting. In routine daily movements, however, it is to be expected that signal artefacts may be induced by

TABLE II
LINEAR REGRESSION VALUES FOR EMG DATA OF ALL PARTICIPANTS

Type	Flexor Muscles			Extensor Muscles		
	Gradient	R^2	r	Gradient	R^2	r
Emb.	4.75×10^{-6}	0.442	1.72×10^{-7}	4.07×10^{-6}	0.290	2.44×10^{-7}
Gel	4.31×10^{-6}	0.461	1.30×10^{-7}	5.28×10^{-6}	0.372	2.82×10^{-7}

[1] Kgf (Kilogram-Force) defined here as equivalent force for a certain weight value, i.e. 1kgf = 9.8N. Therefore, the force values in Newton are 49N, 73.5N, 98N, 122.5N, 148N and 171.5N respectively.

movement of the fabric with respect to the skin, especially where movements are dynamic and non-isometric. The effect of this depends mainly on the fabric and type of clothing into which the electrodes are integrated, rather than the electrodes themselves, since this is what will affect the consistency of the connection between the electrodes and the skin. While a full analysis of the latter is out of the scope of this work, the next section provides some preliminary experiments assessing the suitability of the electrodes for measuring and analyzing muscle activity in dynamic movement tasks.

V. Proof of Concept

This paper describes a methodological design process for the creation of embroidered electrodes for wearable sEMG measurements. As a proof of concept to be tested in more dynamic environments, the resulting electrodes were integrated into jogging leggings.

A. *Implementation*

The set of design variables validated in this study were used to make the embroidered electrodes sewn inside a pair of leggings. Penny-sized flexible printed circuit boards (PCBs – refer to Fig. 3) were made for sEMG acquisition, with an additional stage for signal rectification. The connections between the PCBs electrodes, batteries and an Arduino board recording data on an SD card were made using the same conductive thread as that used for the embroidered electrodes. A zigzag pattern is used for the connection lines. The steel thread has a resistance per length of 91.8Ω/m whereas for copper this is 0.0098Ω/m. The patterns are therefore embroidered for 3 iterations to reduce the effect of thread resistance through paralleling. Fig. 8 shows the interior and exterior view of the leggings as well as a participant wearing them.

B. *Experiments*

A limited set of experiments with N=2 participants (mean weight 75±2.83kg, and mean height 177±5.66cm) are performed as a proof of concept within the scope of this paper. For each participant, data is collected from three running trials (each of 5km in length) on sand, asphalt, and an athletics track located in London's Hyde Park, where participants run at their normal training speed. Note, there is a minimum 24 h resting period between trials, to allow for muscles to recover.

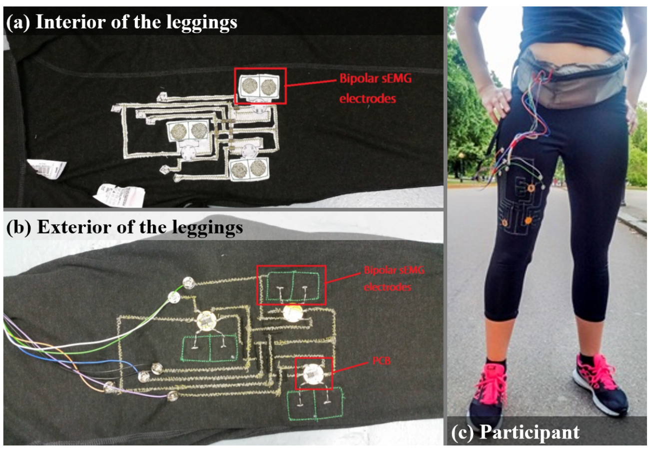

Fig. 8. Proof of concept leggings made using the results of this study. (a) Interior view, (b) Exterior view and (c) worn by a participant in the experiment.

TABLE III
Percentage increase in iARV

Muscle	Asphalt	Sand	Athletics Track
Vastus Medialis	100.04%	127.71%	54.9%
Rectus Femoris	100.02%	126.75%	121.22%
Vastus Lateralis	99.14%	100.07%	35.9%

In amplitude based analysis, the presence of muscle fatigue increases the instantaneous value of the EMG signal with time [25]. To quantify muscle fatigue, a signal analysis script written in MATLAB applies the instantaneous average rectified value (iARV). Table 3 summarizes the percentage increase of the iARV signal for each muscle whilst running on each surface for participant 1. It can be seen that running on the sand surface leads to the maximum amplitude increment of the iARV signal for the three muscles whereas the athletics track led to the minimum increase. Similar results are observed for participant 2, indicating that running on sand increases the likelihood of suffering from fatigue when compared to asphalt or athletics track surfaces confirming the expected behavior of the muscles. These experiments were performed as a proof of concept and, as the number of participants are limited, are not statistically reliable to form a meaningful conclusion on how different surfaces affect muscle fatigue. They do however show the effectiveness of the design approach taken in this work in producing a wearable muscle monitoring solution.

VI. Conclusions

This paper provides for the first time a thorough analysis of electrical characteristics and behavior of embroidered sEMG systems. Through this process, anomalies due to manufacturing inaccuracies of the digital embroidery machine at low dimensions were identified, that lead to results not predicted by theoretical analysis. These need to be considered during the design of embroidered electrodes. The embroidered sEMG system was tested with force-EMG relationship experiments proving the feasibility of its use in acquiring measurements of muscle activity.

As a proof of concept, the embroidered electrodes were embedded in jogging leggings and tested on 2 participants in different running tracks available at London's Hyde Park, looking at how different types of surfaces affect muscle fatigue. The experiments presented here show that wearable sEMG technology is feasible and that it has real applications. The wearable sEMG device is cheap, robust, easy to build and suitable for the everyday user.

These results enable future research in the area that can lead to a wearable garment with integrated sEMG sensors. Such a garment can serve as a low-cost and easy to use interface for its wearer to interact with computers or robots, especially for people with disabilities. For robotic prosthetics in particular, textile sEMG would lead to easy integration with daily life without concerns on sensor placement accuracy. Using embroidery rather than woven or knitted fabric, enables more customization in the design and shape of the electrodes and thus the ability to adapt to different application and monitoring scenarios.

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