Wearable Textile-Based Device for Human Lower-Limbs Kinematics and Muscle Activity Sensing

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Abstract. Lower-limbs kinematics and muscle electrical activity are typically adopted as feedback during rehabilitation sessions or athletes training to provide patients' progress evaluation or athletic performance information. However, the complexity of motion tracking and surface electromyography (sEMG) systems limits the use of such technologies to laboratory settings and requires special training and expertise to carry out accurate measurements. This paper presents a new wearable textile-based muscle activity and motion sensing device for human lower-limbs, which is capable of recording and wirelessly transmitting sEMG data for several specific muscles as well as kinematic parameters, allowing outdoor and at-home use without direct supervision by non-expert users. In particular, this work is focused on the development and analysis of textile electrodes and garment design, as well as the definition of a proof-of-concept study for sEMG data recording. Obtained values were compared against average rectified values (ARV) recorded using a gold-standard conventional wireless sEMG system. Apart from one muscle (vastus medialis), the developed device showed overall promising results in the muscle activity sensing for lower-limbs, highlighting its possible use in the rehabilitation and sport performance fields. In addition, a washing test was conducted on the electrodes, where it was shown that the proposed textile electrodes maintained structural integrity and showed an acceptable level of electrical parameters deterioration when comparing pre and post washing characteristics.

Keywords: Textile · Electrodes · Electromyography · Wearable · Smart Garments

1 Introduction

In the modern world, wearable devices permeate deeper and deeper into our lives. Personal health care and well-being are some of the fastest-growing areas for consumer electronics where these technologies are increasingly used. This can be seen by the pervasive adoption of next-generation smart watches, activity trackers and wearable heart rate monitors. The overall rise of connected devices is forecasted to increase from 593 million in 2018 to 929 million devices by 2021 [1].

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The current trend in the development of wearables relies on the technical advancements in systems integration. Up-to-date devices need to be discreet, easy-to-use, unobtrusive and able to perform continuous and remote monitoring in real time. Even though implantable devices seem to meet mentioned requirements, their development still poses numerous technical, medical and security issues [2, 3], making their use currently unfeasible in practice.

An alternative approach involves the integration of wirelessly connected technologies into items frequently used in everyday life (such as clothing). Thus, smart garments represent a potential solution for this problem. Research and development in this area in recent years is gaining wide popularity, as can be seen by the body of literature in the space, and by the large number of funded projects and an exponential increase in the number of associated publications [4].

One of the technologies whose development in this direction seems promising is surface electromyography (sEMG). Monitoring of muscle activity can be extremely useful in sports and medicine, especially physiotherapy and rehabilitation. However, the implementation of long-term regular monitoring has proven to be challenging using traditional sensing solutions (in particular, standard self-adhesive pre-gelled electrodes), due to these being relatively uncomfortable and possibly causing skin irritation and contact allergic dermatitis [5, 6].

Furthermore, the majority of commonly used devices utilize connecting copper wires which can restrict movement and cause motion artifacts in the recorded signal. Moreover the construction of adhesive electrodes makes the direct embedding of these into garments virtually impossible. An alternative solution for a wearable sensing solution which is compatible with daily regular use can be textile electrodes, which can be easily embedded into clothing and are comfortable and safe for long periods of use. In this work, the authors present an implementation of this concept: wearable device – smart leggings, which contain embedded textile electrodes as well as inertial sensors to measure lower-limbs kinematics. The following work will describe the technical process put in place for the development of the smart garments, as well as the functional performance testing procedure carried out for evaluation.

The manuscript is organized as follows: related work and current design challenges are briefly described in Sect. 2, while Sect. 3 is dedicated to addressing those issues in textile electrodes and garment design. In Sect. 4 descriptions and results of the washing test and proof-of-concept study are presented. Finally, Sect. 5 contains a discussion of the obtained results, overall conclusions and future perspectives.

2 Related Work

Several works investigating the development of smart garments with built-in sEMG for lower-limbs monitoring have been presented in recent years.

Catarino et al. [5] designed a swimsuit for monitoring biometric and performance parameters of athletes. The base of the suit is polyamide and elastane yarns using seamless knitting jacquard machine, while electrodes and connective paths are simultaneously knitted with conductive yarn. In other works, the authors used the same technology for manufacturing e-leggings with built-in sEMG functions [7]. Electrodes were knitted with silver-coated multifilament yarn and placed to record the electrical activity of the following muscles: vastus intermedius, rectus femoris, biceps femoris, tibialis anterior and gastrocnemius medialis. Electrodes placement and inter-electrode distance were chosen according to SENIAM recommendations. Authors also investigated two types of yarn, the Elitex, conductive yarn made of polyamide fibres with silver coating and Bekitex Mn 50/1 made of polyester and stainless steel. According to the authors, Elitex yarn showed better impedance stability under strain [8]. Despite the relatively low signal-to-noise ratio (SNR) values for knitted electrodes, the authors showed that it is possible to successfully register the electrical activity of muscles using knitted electrodes.

Jogging leggings with embroidered electrodes for recording quadriceps muscle electrical activity were presented by Manero et al. [9]. Stainless-steel thread-based (Sparkfun DEV-11791) sEMG electrodes were made using an embroidery machine and placed on pair of mass-produced jogging leggings. The areas underneath electrodes were thickened with an additional layer of felt fabric. Electrodes were placed according to SENIAM recommendations with 1 cm inter-electrode distance in unstretched state. The authors stated that this device was able to record the difference in muscle performance when running on various surfaces, such as sand, asphalt, and athletic track.

A prototype for recording upper leg muscle groups via sEMG was presented in [10], with authors using sewn-on textile electrodes, even though the type of fabric used was not disclosed in the work. Authors investigated agreement in average rectified values (ARV) of sEMG for their device and traditional Ag/AgCl electrodes and found it to be within 95%. This prototype was subsequently modified by the developers in the development of a commercially available product, e.g. the compression shorts Mbody by Myotec [11] that record cumulative sEMG data from different muscle groups: quadriceps, hamstrings, and gluteus. A similar device currently on the market is Athos [12], which is a wearable system with sEMG electrodes integrated into compression athletic apparel for the upper and lower body. Unlike Mbody, Athos compression shorts include separate sensors for outer quadriceps, inner quadriceps, hamstrings and gluteal muscles. Athos's electrodes are composed of conductive polymer ink applied to the fabric surface. A similar device was announced by B10nix (B10NIX Ltd., Milan, Italy) [13]. However, it is only available for pre-order and, to the best of the authors' knowledge, no information considering design or validation was disclosed.

Alternatively, beside the shorts and leggings solutions described, the implementation challenges associated with the development of a prototype of smart socks was also investigated in [14]. This device detects electrical activity of the gastrocnemius and tibialis muscles. Non-adhesive hybrid polymer electrolytes-based electrodes (polyvinyl alcohol and carboxymethyl cellulose blend complexed with 30 wt. % of NH_4NO_3) were used. In the case study, smart socks were used to detect the risk of falling.

While some promising results in wearable muscle activity tracking devices development have been shown, some shortcoming of smart garments are still present, such as motion artifacts, effects of fabric and electrodes stretching, unstable skin-electrode impedance, high cost for commercially available models, limited washability and, therefore, device lifetime. Those issues will be discussed and addressed in the presented solution in the following sections.

3 Implementation

A variety of methods have been considered as options for the development of the proposed textile electrodes with the ultimate goal being to ensure the suitable electrical properties and washability of appropriate wearable sensing systems. This includes for example poly (3,4-ethylenedioxythiophene): polystyrene sulfonate (PEDOT: PPS) electrodes that are actively gaining popularity recently due to their biocompatibility properties, as well as their electrochemical and thermal stability [15]. Such electrodes can be produced using a wide variety of methods, i.e., by soaking fabric in polymer solutions with subsequent treatment (drying, heating etc.), using screen printing or inkjet printing techniques. However, these electrodes need further improvement since they currently show inferior results in terms of washability and electrical parameters than textile electrodes made of silver-coated fibres [16]. Such textiles, woven, non-woven and knitted, are also widely available and inexpensive. While the electrical parameters of knitted conductive textile are unstable when worn due to stretching of the fabric [17], woven fabric, such as nylon ripstop, show better performance characteristics than knitted and non-woven [18], possibly due to tighter weave, relatively even surface of the fabric, and subsequent increase in the skin-electrode contact area. Thus, commercially available nylon silver-coated woven fabric Bremen RS (Statex, Bremen, Germany) was used for the textile electrodes. This fabric is suitable for medical applications and shows promising electrical properties [19].

The electrode-skin contact impedance is a major factor in obtaining good quality sEMG signals and this is highly dependent on the skin hydration levels; even though moistened textile electrodes proved to be comparable to traditional electrodes in terms of impedance and recording quality, however, drying of the electrode surface over time significantly worsens their electrical performance [20]. In practice, a sufficient level of moisture can be ensured by accumulating skin perspiration under electrodes over time and avoiding evaporation by using a polymer coating on the back of the textile electrode [21]. To avoid possible degradation of the conductive fabric electrical properties, instead of coating the back of the electrode with polymer solution, a thin polymer sheet was glued onto the base fabric, with a larger piece of conductive fabric placed on top (Fig. 1a), with the result that only the edges of the conductive fabric were glued to the base fabric and most of the electrode surface had a waterproof layer underneath.

Another technical challenge in the development of smart garments is ensuring robust, yet flexible electrical connections between the electrodes and associated integrated electronics. Thin multi-stranded wires with reinforcement thread were attached to the electrodes by sewing them with a conductive thread, with a technique similar to the one used in [22]. The wire core was twisted into a small loop and secured in place by 6–8 stitches, and afterwards stitches were placed along the edge of the conductive fabric and back to the loop (Fig. 1b).

This technique made the realization of a flexible connection both durable under mechanical strain and washable. For the current prototype, stitching was performed manually; however, sewing or embroidery machine can be also used in a manufacturing process. Manufactured textile electrodes shown in Fig. 2.

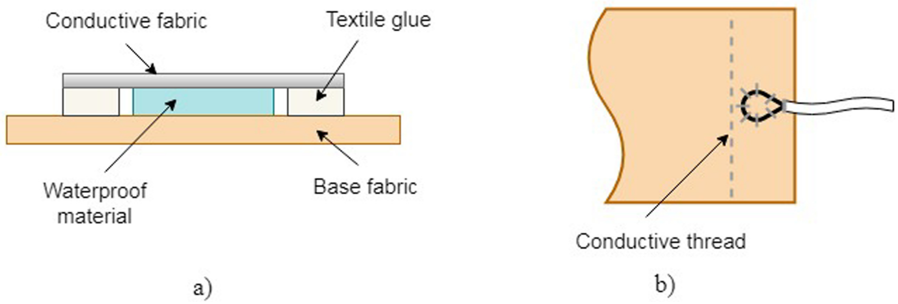

Fig. 1. Construction of textile electrodes

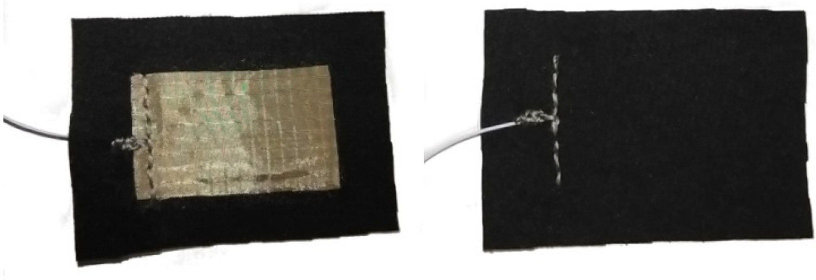

Fig. 2. Textile electrodes: front and back view

Electrodes were sewn together in pairs with 2 cm inter-electrode distance and were additionally strengthen with several lines of machine stitching to prevent fabric stretching during movements, since it can result in changes in inter-electrode distance, affecting registered signal. Repetitive stretching can also lead to deterioration of glue seams and conductive fabric frying. It was previously reported in [23, 24] that soft padding can help reduce motion artifacts; therefore, an additional level of felt was added to the electrode pairs. Each pair was placed according to SENIAM guidelines on a base inner fabric for the following muscles: rectus femoris (RF), vastus medialis (VM), biceps femoris (BF), semitendinosus (ST), gastrocnemius medialis (GM) and lateralis (GL). The first four muscles are on the thigh, while the last two are on the calf. Connective wires were threaded through the fabric from the inner layer to improve user comfort and were attached to the fabric using zigzag stitches into respective places. A spare length was left to allow free stretching of fabric when worn (Fig. 3).

To simplify the next step of the prototyping process, mass-produced sports leggings were used as an outer layer. Light compression provided by such garment also ensures desired pressure on electrodes and the overall fit of the garment. Signals from the electrodes are transmitted to two electronic units, a slave unit for the calf muscles and a master unit for the thigh muscles. Registered sEMG then transformed into ARV envelopes by smoothing amplified and rectified signal with low pass filter. Using ARV allows lowering the sampling frequency and present sEMG signals in the conventional form [25].

Fig. 3. Electrodes placement on the base fabric: front and back view

The device also obtains the body segments (lower and upper leg) orientation and transmit the collected data wirelessly to a smartphone, based on the system architecture described in [26]. The slave/master units' holders are attached to the outer layer fabric. The units can be easily removed from holders, thus leaving the leggings without any active electronics when needed to be washed. A general view of the device is shown in Fig. 4.

Fig. 4. General view of the designed device

4 Performance Tests

4.1 Proof-of-Concept Study

For this proof-of-concept, only two volunteers were recruited: female, height - 164, 167 cm, weight - 59.5, 56 kg, respectively. Each subject was informed about the nature of study, possible risks and the tasks given. Each subject completed the consent form and the Physical Activity Readiness Questionnaire (PAR-Q) and had no self-reported musculoskeletal and skin injuries or disorders. Clinical Research Ethics committee of the Cork Teaching hospitals approval was obtained to evaluate the device on human subjects.

Volunteers were initially asked to perform a set of exercises while wearing the designed device. Exercises were chosen in order to ensure the activation of separate muscle groups (Table 1). Participants wore the leggings under evaluation, and a 5-min pause was taken before the start of the experiment. No additional skin preparation was carried out. After 10 min rest, the same set of exercises was replicated in the same order with a gold-standard sEMG system (BTS FREEEMG, BTS Bioengineering, Italy with standard pre-gelled 24 mm adhesive electrodes from Covidien Kendall).

Table 1. Performed exercises

Exercise	Major muscles engaged	No. repetitions
Sitting knee extension	Vastus medialis, rectus femoris	5
Standing hip flexion/extension	Rectus femoris, vastus medialis/bicepsfemoris, semitendinosus	5
Plantar flexion (standing on tip-toes)	Gastrocnemius medialis/lateralis	5

MATLAB software was used to process all gathered data. The ARV of the sEMG activity signal was extrapolated. Raw data were exported from BTS EMG Analyzer and then rectified and averaged using a moving average by sliding a 200 ms window to obtain the ARV records, as recommended in [25, 27].

It is important to note that, due to the impossibility of placing the electrodes from the two systems simultaneously on the same muscles, the compared signals were recorded in different sessions, therefore differences between muscles activities evaluated in these initial trials can naturally occur. Despite this possibility, results obtained during the experiments are promising as the muscle activation times and sequences show a similar pattern to those obtained with the gold-standard system (Figs. 5, 6, 7 and 8). The amplitude of the obtained sEMG signals are also comparable, considering that the analog circuit of the wearable device contains an additional amplifier (factor of 10). For gastrocnemius muscles, however, current pre-set amplification of the signal led to saturation and a partial loss of the signal (Fig. 8). This issue does not affect the electrodes design and the circuit gain can be suitably adjusted in future trials.

Simultaneous recording of all major leg muscle made it possible to analyze the influence of motion on the quality of signal for each electrode pair. The VM muscle electrodes showed unsatisfactory results. The signal from this muscle was repeatedly lost. The proximity of the electrodes to the knee, when bending, results in an electrode displacement and consequent loss of the skin-electrode contact.

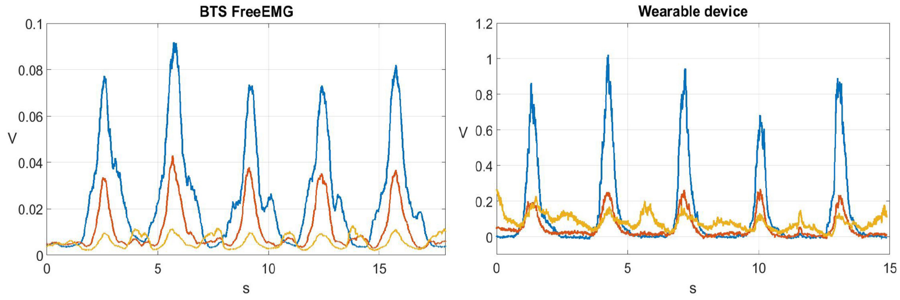

Fig. 5. Sitting knee extension: blue – RF, red – BF, yellow – ST

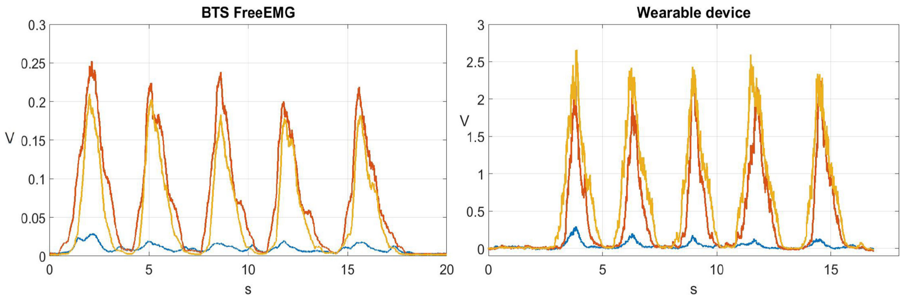

Fig. 6. Standing hip extension: blue – RF, red – BF, yellow – ST

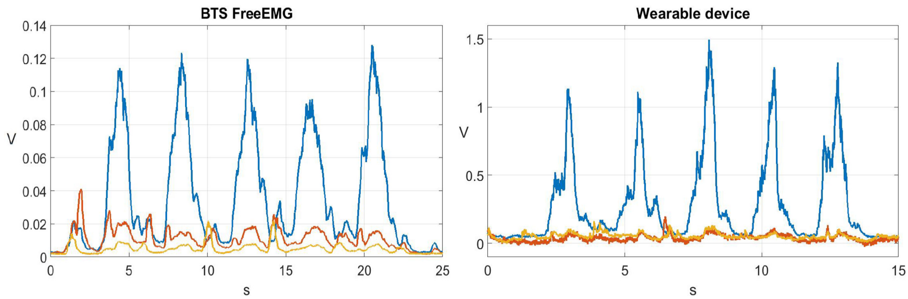

Fig. 7. Standing hip flexion: blue – RF, red – BF, yellow – ST

4.2 Washing Test

To assess the washability of the developed electrodes (shown in Fig. 1), a sample of 10 electrodes underwent 20 manual washing cycles. Five ml of mild detergent (liquid detergent for delicate fabrics) were diluted in 2 L of warm water at 33.8 °C (SD: 1.5 °C) as recommended by the manufacturer. Electrodes were allowed to soak for three minutes and then were gently “swished” through the water. Electrodes were then thoroughly rinsed three times in clean warm water 34.4 °C (SD: 1.2 °C) and placed flat on a thick

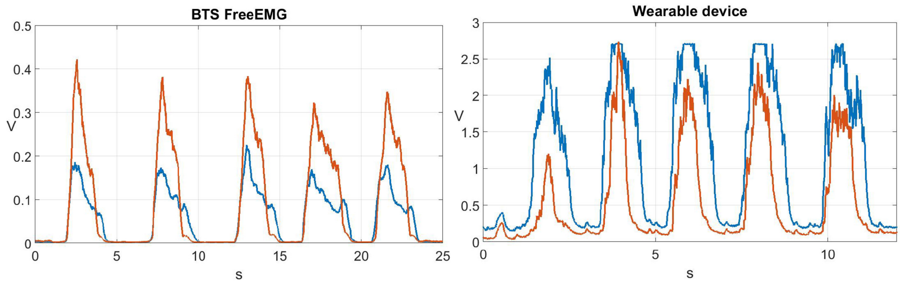

Fig. 8. Plantar flexion: blue – GM, red – GL

towel and gently pressed with another towel to absorb the water in excess. Electrodes were finally placed flat on a rack in a well-ventilated room until dry.

To characterize the electrical properties of the electrodes, the resistance in the path from the end of the 12 cm wire to the central area of the electrode surface was measured using a digital multimeter HP 34401A. Resistance measurement was taken 10 times for each electrode and averaged. Electrodes resistance was measured before and after 5, 10, and 20 washing cycles. Obtained values are presented in Table 2.

Table 2. Electrodes resistance (Ω, SD)

Electrode	Washing cycle			
	Before	5	10	20
1	1.29 (0.18)	1.71 (0.21)	2.83 (0.13)	4.10 (0.28)
2	1.33 (0.13)	1.25 (0.02)	2.74 (0.32)	2.49 (0.36)
3	1.37 (0.17)	0.91 (0.04)	2.58 (0.32)	2.88 (0.26)
4	1.32 (0.11)	1.61 (0.07)	2.91 (0.14)	2.97 (0.36)
5	1.46 (0.25)	1.62 (0.12)	1.71 (0.13)	2.33 (0.14)
6	1.58 (0.15)	1.78 (0.06)	2.26 (0.22)	2.19 (0.07)
7	1.28 (0.03)	1.24 (0.02)	1.40 (0.09)	1.70 (0.03)
8	1.26 (0.13)	0.93 (0.05)	1.78 (0.07)	1.57 (0.09)
9	1.29 (0.24)	1.62 (0.09)	2.69 (0.19)	3.18 (0.25)
10	1.85 (0.34)	4.11 (0.49)	5.05 (0.76)	–

Textile electrodes retained their performance throughout all 20 washing cycles. Figure 9 shows the sEMG obtained using washed electrodes. The observed increase in resistance after 20 cycles has not exceeded 3 Ω and averaged at 1.24 Ω.

Sewn connections with conductive wires remained fully intact, apart from one electrode (10), most likely indicating an inconsistency in the manually performed wire insulation removal. Glued parts of conductive fabric started to peel off for a few electrodes,

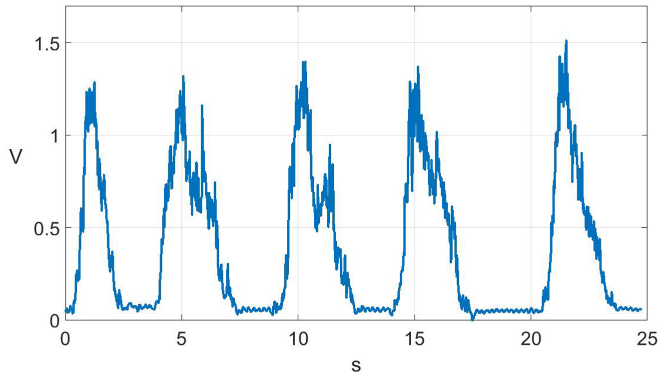

Fig. 9. Plantar flexion (GL), electrodes after 20 washing cycles

resulting in the slight fraying of the edges. This issue was observed in electrodes manufactured in one batch and while this has not affected the overall electrical properties of electrodes, this is a problem to be addressed in the future.

5 Conclusion

Wearable devices with embedded sEMG can open new opportunities in the field of rehabilitation and professional sports, which include a collection of data in real-world conditions, such as home-settings or athletes' outdoor practices. Ease-of-use of wearable devices can ensure that potential users will not need any training or previous knowledge related to sEMG, which would allow remote monitoring of patients and athletes without direct supervision from a medical professional in rehabilitation. The prototype of such a device was developed and presented in this paper and its potential applicability was evaluated in the proof-of-concept study and in the functional performance test. Furthermore, methodologies adopted for textile electrodes manufacturing were described in detail.

Preliminary results show that the presented device is capable of registering sEMG in form of ARV and obtained results are comparable to a gold-standard system. The study confirms previous findings that wearable sEMG technology is feasible and promising in research, medical, and sports applications. The washing test, conducted on a sample of electrodes, showed that conductive fabric and electrode-wire connections maintained their properties during 20 washing cycles. However further investigation of the gluing methods of conductive fabric to the base fabric is still needed. A possible solution to prevent frying of conductive fabric might be replacing adhesive with sewn connections entirely.

While the problem of motion artifacts was reported previously in the literature, the results of this study show that only VM signal was significantly affected during exercise performance. This information can provide insight into improvement of existing engineering solutions to ensure reliability for every muscle of interest, possible options include altering the sewing pattern of the leggings and/or the use of anti-slip materials. Further investigation on a bigger sample could shed light on the less prominent impact of motion artifacts on skin-electrode impedance. Long-term use is also a point of further investigation. Follow-up studies with sufficient sample size are planned to evaluate the validity and reliability of the proposed device and method for sEMG monitoring. Moreover, prototypes of various sizes need to be produced to reflect the diversity of the population and avoid possible bias in future trials.

The possibility to integrate the developed sEMG monitoring system in a real-time functional electrical stimulation system for feedback control is also a matter for future studies.

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