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Characterization of embroidered textile-based electrode for EMG smart wear according to stitch technique

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Abstract

This study fabricated and evaluated the textile-type electrodes for application to smartwear that can measure surface electromyography(sEMG). It was manufactured by lock stitch(LS) and moss stitch(MS), and the stitch distance was prepared as 1, 2, or 3 mm. The surface and compression property was measured by using the Kawabata evaluation system, and the sheet resistance and skin-electrode impedance were analyzed. The coefficient of friction(MIU) of the MS was larger than that of the LS. On the other hand, the geometrical roughness(SMD) showed a smaller value. When the same load was applied, the compressive range of the MS was larger than the LS. When it was manufactured as a leg sleeve and worn, the conductive path could be increased as the loops made of conductive yarn become flat as the loops adhere to the skin by the pressure of clothing. Accordingly, the skin-electrode impedance decreased by increasing the area in contact with the skin. As the results of the RMS(root-mean-square), the LS was higher than the MS in a stable. Nevertheless, the SNR(signal-to-noise ratio) value was lower than that of the MS because movement generated noise during operation. Therefore, more stable signal acquisition is possible when applying MS. It is expected that could be applied to producing smartwear for sEMG measurements with superior sEMG signal acquisition performance while having a softer touch and flexibility.

Keywords: Textile electrode, Surface electromyography, Smartwear, Embroidery, Stitch Technique

Introduction

Recently, as the interest in wellness and health has increased, research and development of wearable devices for monitoring biosignals and muscle activation are increasing rapidly (Ahsan et al., 2022; Zahid et al., 2022). Surface electromyography (sEMG) signals are used mainly to analyze muscle activity that is used through the electrical potential of the human body that changes during daily life or exercise (Lam et al., 2022; Zheng et al., 2022). Accordingly, human bioelectrical signals are collected through wet and dry electrodes to detect and continuously monitor them. The electrode traditionally used to measure sEMG is the silver/silver chloride (Ag/AgCl) electrode, which has the advantage of low skin–electrode impedance. This electrode is also disposable and easy to use. However, there is a limit when applied as a wearable device or smart wear because of issues, such as deterioration of the adhesion performance and signal collection performance caused by long-term use and skin. In addition, for sEMG measurements, electrodes must be attached to the corresponding muscle, and it is difficult to measure the muscle in a location that is not reachable or invisible. When applied in the form of clothes, it is not possible to attach the wet-electrode long-term, and it is expensive compared to clothes worn in daily life. Moreover, there are limitations, such as discomfort when wearing. Therefore, sEMG-based smart wearable systems are being developed in clothes, sleeves, and protectors according to the purpose (Choudhry et al., 2021).

Many previous studies have developed a textile-type dry electrode to replace the existing Ag/AgCl hydrogel electrode to overcome the limitation of smart wear for sEMG measurement. In general, as a method for manufacturing textile-type dry electrodes, there are methods of using a conductive yarn in weaving or knitting form, a decoration form on fabric using an embroidery technique, and a method of coating or laminating a substrate fabric or yarn on a conductive material (Choi & Hong, 2019; Kim et al., 2022b; Ohiri, et al., 2022; Shuvo et al., 2022; Zhao et al., 2022). Among them, embroidery is one of the most often used technologies in the sphere of smart textiles. It provides a great asset in wearable and flexible electronics development, such as bio-sensing sensors (Mecnika et al., 2014). Embroidery is a patterning method through which a series of overlapping stitches is formed from a single continuous thread, or multiple threads, on a finished fabric substrate. This method is one of the earliest for circuit routing and interconnections in e-textile fabric systems. Moreover, it has been used in recent times for electrode structures. By producing several intersections of a continuous thread, greater conductivity can be imparted to the embroidered structure compared to a single conductive thread (Katherine et al., 2022). Textile electrodes manufactured using embroidery can collect internal and external signals from the inside and outside based on the fabric due to the loop. In addition, embroidered electrodes can provide decorative conductive patterns on a finished textile surface (Goncy-Berk & Tuna, 2021). Generally, the most commonly used embroidery stitch techniques include lock and moss stitches. The lock stitch is a method that skips one stitch and proceeds in a linear type of two-dimensional form, like the existing sewing seam. The technique is performed by moving the needle up and down and fixing the upper thread to the fabric by the lower thread wound on the bobbin. These techniques have been used to produce electrical circuits and antennas. The moss stitch makes a loop shape by pulling up a thread wound on a bobbin using a hook-type needle. This type forms a three-dimensional structure that can be used to manufacture a textile sensor with direct contact by controlling the moss height (Kim et al., 2022b; Rho et al., 2022).

In a previous study, to apply to EMG smart wear for fitness, the characteristics of embroidery electrodes were used to manufacture each shape using a lock-stitch; a basic embroidery stitch technique and optimal conditions were derived (Kim et al., 2022a). In the case of conductive yarn, however, it is rigid and stiff because the surface is coated with metal. Therefore, when the sensor or electrode is made with a lock stitch, it is rough and hard. Thus, it has a disadvantage in that the flexibility and touch are poor. In previous studies, the comfort of the conductive fabric for application to smart clothing is as much an important factor as the performance (Lee, 2022; Tadesse et al., 2018; Zhao et al., 2022). Tadesse et al., (2018) measured the tactile comfort of a conductive fabric using Kawabata's evaluation system and then compared the results with those of conventional textile products. The results showed that polyester knitted fabrics incorporated with copper were rougher than pure polyester fabric. Lee, (2022) reported the relationship between the functionality and comfort of conductive flat-knitted fabric for smart clothing. The surface roughness of the course direction increases as the blending ratio of the conductive yarn increases. In addition, the recoverability from compression decreases as the blending ratio of conductive yarn increases, and the work of compression increases. This suggests that the conductive fabric can be compressed easily but is less likely to recover from compression. Zhao et al. (2022) proposed the 3D textile electrode inspired by the embroidery technique with a 1D composite yarns containing a mixture of reduced graphene oxide (rGO), sericin, and water-retention polymer. The 3D knitting porous construction comforted users because of the soft tactile formation and the breathable property. In addition, it was confirmed that the contact impedance was lower than that of the reference electrode. They reported that accurate measurement was possible by analyzing the muscle fatigue process.

Accordingly, research on the tactile comfort, electrical, and bio-signal acquirement performance of conductive weaves or knits is continuously being conducted, but the embroidery technique is relatively insufficient. Our research team confirmed that EMG signal collection was possible with the lock-stitch technique, but since it was densely stitched with rigid conductive yarn, the surface of textile electrode were rough. Accordingly a moss stitch with a different structure made by loop structure was added to compare the difference in tactility and EMG signal collection performance between both stitch techniques. Therefore, in this study, an electrode capable of realizing a more flexible and soft touch was manufactured by applying a moss stitch that forms a three-dimensional structure while the yarn forms a loop. The fabricated electrode was compared and evaluated for the tactile, electrical, and EMG collection performance with the lock-stitch embroidered electrode developed in a previous study.

Experimental

Materials

The substrate fabric used to manufacture the embroidery electrode was a polyester-based elastic fabric for application as smart sportswear. The fabric was composed of 88% Polyester and 12% Spandex, and the weight and thickness were 451.5 g/m^2 and 1.0 mm, respectively. The substrate fabric used was a double weave (wale $\times$ course, 128 $\times$ 88/5 cm), and the face of the fabric is composed of flat knitted fabrics, and the backside is double-side knitted fabrics. This study used the same materials reported elsewhere (Kim et al., 2022a). The designed electrodes were manufactured using a technical embroidery machine (SGVA 0109-825, ZSK Stickmaschinen GmbH, Krefeld, Germany). The needle used for the embroidery machine was No. 14 with 90 Nm. At this time, the conductive yarn, silver coated polyamide/polyester hybrid thread, from the AMANN group (Silver-tech, AMANN group, Bönnigheim, Germany) which has a 28 Tex and resistivity of 530 Ω/m, was used for the upper and lower yarns.

Design and preparation of embroidery textile electrodes

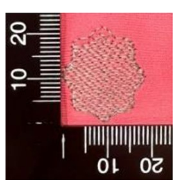
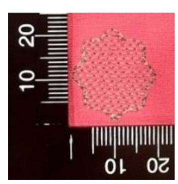
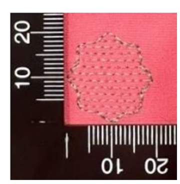
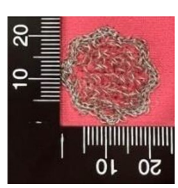
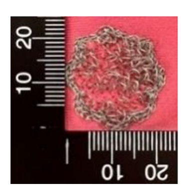
As the shape of the electrode for the EMG acquirement, the design was carried out by selecting the fill (WF) of the wave-type with a diameter of 20 mm, which was derived as the optimal design for EMG acquirement in previous studies (Kim et al., 2022a). The samples were designed using the EPC_win embroidery design program (ZSK Stickmaschinen GmbH, Germany). Two stitch techniques were selected: a lock stitch (LS) manufactured in a 2D form and a moss stitch (MS) manufactured in a 3D form. Both stitch techniques applied the same embroidery parameters of a stitch length of 2.0 mm. The MS type electrode was manufactured by selecting a loop height of 3 mm through a previous study (Rho et al., 2022). The change was analyzed according to the density of LS and MS by designing the stitch distances as 1 mm, 2 mm, or 3 mm (Fig. 1).

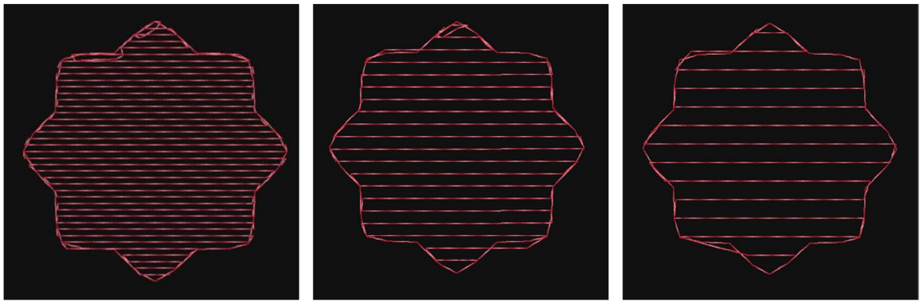
A bipolar-type electrode is required to acquire an EMG signal. Accordingly, in this study, two identical electrodes were designed with a distance between their centers (IED, inter-electrode distance) equal to 40 mm, as reported elsewhere (Kim et al., 2020, 2022a). The designed samples were manufactured with a technical embroidery machine using silver-based conductive yarn on the substrate fabric. Table 1 lists the images and sample codes of the manufactured samples.

Design and preparation of leg sleeves embedding the embroidery textile electrodes

This study targeted the rectus femoris muscle. The leg sleeve embedded with embroidery textile electrodes was fabricated for surface electromyography (sEMG) (Fig. 2). sEMG was measured for one subject (sex = female, age = 32, height = 175 cm, body weight = 60 kg). According to the subject's condition, the length was 180 mm based

Table 1 Sample codes and images of the samples in this study

Stitch technique	**Lock stitch**		
Sample code	LS_1	LS_2	LS_3
Stitch distance (mm)	1	2	3
Sample image			
Stitch technique	**Moss stitch**		
Sample code	MS_1	MS_2	MS_3
Stitch distance (mm)	1	2	3
Sample image			



(a) 1 mm **(b)** 2 mm **(c)** 3 mm

Fig. 1 Image of the embroidery textile-based electrode design according to the stitch distances (**a**) 1 mm, (**b**) 2 mm, and (**c**) 3 mm

on the electrode part, and the circumferences of the upper and lower thighs corresponding to the position were measured. This study approved by the Institutional Review Board of Korea Institute of Industrial Technology (KITECH) (IRB No. 2022-002-001 22 June 2022). According to previous studies (Kim et al., 2020, 2022a), a 30% pattern reduction rate (PRR) was applied, considering the elasticity of the substrate fabric.

Characterization

Morphology

The morphology of the embroidery textile-based electrodes was analyzed according to the stitch technique and distance by measuring at × 35 magnifications using a multimedia imaging microscope (RH-2000, KEYENCE Co. Ltd., Seongnam, Korea). In addition, to compare with lock and moss stitch, 3D images were taken using Zeiss Xradia 510 Versa 3D X-ray microscopes (XRM, (Carl Zeiss Microscopy Deutschland GmbH, Oberkochen, Germany). The measured voltage and power were 50 kV and 4 W, respectively, and the resolution was 22 µm. The number of measured images was 1601 per sample, and the exposure time was five seconds. The amount of conductive yarn used in the textile electrode was confirmed by calculating and comparing it according to the stitch length, distance, number of stitches, and fabric thickness through the EPC_win

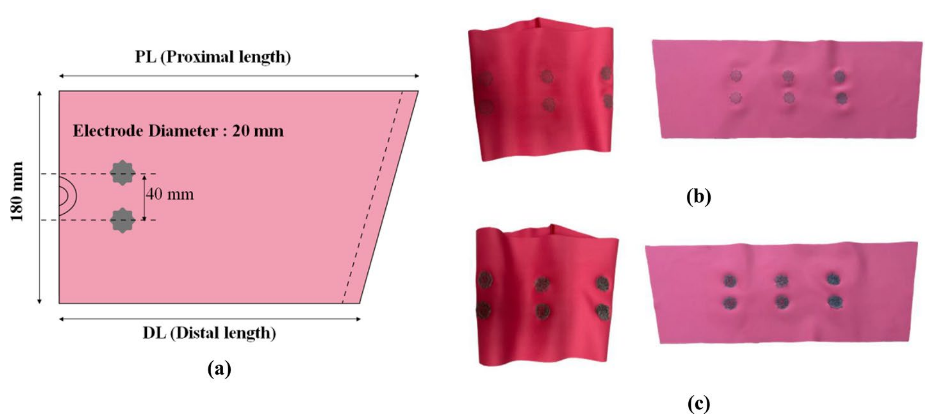


Fig. 2 **a** Schematic diagram of the preparation of leg sleeves with embedded embroider electrode; actual sample image of (**b**) wave fill lock stitch (WF_LS) sample, and (**c**) wave fill moss stitch (WF_MS) sample

Table 2 Measured condition of the samples

Property	Symbol	Characteristic value	Unit	Test condition
Surface test	MIU	Coefficient of friction	–	Tension: 400 g Static friction load: 25 g Roughness static load: 10 g
	MMD	Mean deviation of MIU	–	
	SMD	Geometrical roughness	μm	
Compression test	LC	Compressional linearity	–	Maximum load: 50 gf/cm^2 Velocity: 0.02 mm/s
	WC	Compressional energy	$gf.cm/cm^2$	
	RC	Resilience	%	
	T0	Thickness at 0.5 gf/cm^2	mm	
	TM	Thickness at maximum compressive load	mm	

embroidery design program (ZSK Stickmaschinen GmbH, Germany) based on previous research (Rho et al., 2022).

KES-FB evaluation

The surface and compressional properties were evaluated and compared according to the conditions of the prepared samples using KES-FB-AUTO SYSTEM (KATO Tech Co. Ltd., Japan). The test environment was conducted at the standard air condition (20 ± 2 ℃ and 65 ± 4% R.H.). The sample size was 20 cm × 20 cm, and the items of the surface and compression properties were measured. All the test methods were performed according to KES-FB standard methods, and the mechanical characteristic values calculated through the evaluation are listed in Table 2.

Surface test was measured with KES-FB4 (KATO TECH CO., LTD., Japan), and MIU (coefficient of fabric surface friction), MMD (mean deviation of MIU), and SMD (geometrical roughness) were analyzed. Surface friction and roughness are measured with sensitive contactor, which is made by steel pianowires of 0.5 mm diameter. This contactor is placed on the surface of the specimen with the compressional force of 50 gf. The specimen is moved by a constant velocity of 0.1 cm/s horizontally where the tension of the specimen is kept 20 gf/cm and the contactor is kept its position. Compression test were conducted with KES-FB3 (KATO TECH CO., LTD., Japan). Here, the factor of compression linearity (LC), compressional energy (WC), compression resilience (RC) was measured and Thickness sample (T0) and thickness at maximum compressive load (TM) also measured. The specimen is compressed by two circular plates of steel having 2 cm^2 area. The velocity of the compression is 20 μm/sec and the pressure attains at 50 gf/cm^2.

Sheet resistance

The electrical properties of the embroidery textile-based electrode were measured according to the stitch technique and distance using a 4-point probe conductivity meter (RSD-IG 4-Probe, DASOLENG., Cheongju, Korea). All samples were measured 10 times, and the average value was used.

Skin–electrode impedance

The impedance of the electrodes is directly related to the signal-to-noise (SNR). Accordingly, the electrodes must undergo an impedance validation that supports efficient EMG acquirement performance. In this study, the skin–electrode impedance after wearing a leg sleeve with two embroidery electrodes embedded in parallel was measured according to the embroidery stitch technique and distance. First, an Ag/AgCl electrode was attached to the measuring position of the leg, and impedance was measured to compare with the embroidery textile-based electrode. The impedance of embroidery textile-based electrodes was measured and compared according to the stitch technique and distance after wearing a leg sleeve so that the embroidery electrode could contact human skin. The distance between the center of these two electrodes was kept at 40 mm, and the impedance was tested using an impedance analyzer (ZIVE P2 ELECTROCHEMICAL WORKSTATION, Won-A tech. Co. Ltd., Seoul, Korea). The skin–electrode contact impedance was measured with an AC sinusoidal signal at the 1 to 1000 kHz frequency range. For comparison, the skin–electrode impedance value at 100 Hz was analyzed according to the electrode type, targeting the peak frequency component for EMG. During the experiment, no pre-treatments of the skin, such as shaving or exfoliation, were carried out separately because people would not implement skin preparations when wearing sEMG suits in practical use. All measurements were performed on the same day on the same subject.

sEMG measurement and data processing

Figure 3a shows a schematic diagram of the measuring process for the sEMG signal by BIOPAC system. In this study, the sEMG signals from embroidery textile-based electrodes were collected and analyzed during knee extension after wearing the leg sleeves (Fig. 3b). Based on a previous study (Kim et al., 2020), bipolar EMG records were obtained using textile-based electrodes on the rectus femoris, which is located on the anterior thigh and is involved in knee extension and hip flexion. In this study, the same method used in previous studies was carried out (Kim et al., 2020, 2022a).

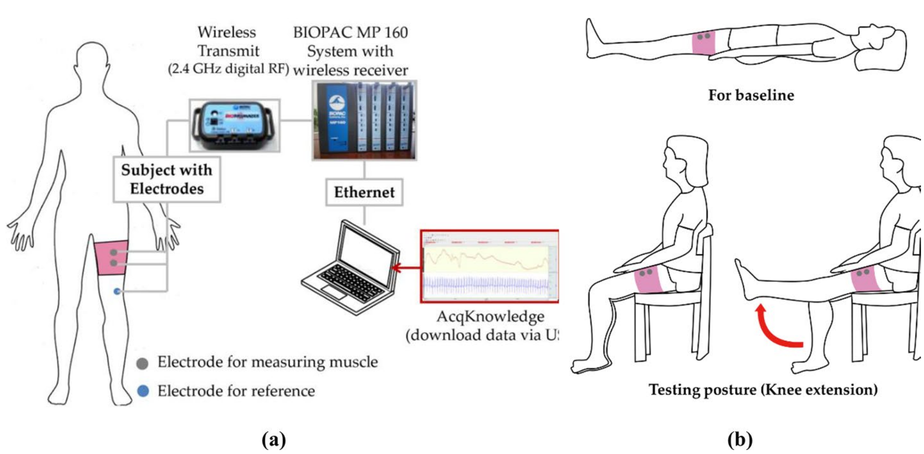


Fig. 3 Schematic diagram for (**a**) measuring using the sEMG signal and (**b**) the testing posture for baseline and knee extension

Before the test, with no skin preparation, the location of the corresponding rectus femoris, which is the position for measurement, was marked on the skin. Both electrodes were positioned on the midpoint of the muscle for the rectus femoris, and it was aligned on the longitudinal axis of the target muscle. A reference electrode, pre-gelled self-adhesive Ag/AgCl electrodes (Kendall LTP, Covidien, MA, USA), was placed over the head of the fibula, the electrically neutral bony prominence. The sEMG signal recordings were amplified and filtered (20–500 Hz) (MP160, BIOPAC Systems, Inc., Goleta, CA, USA). The entire data processing of sEMG was performed using AcqKnowledge 5.0.1 Software (BIOPAC Systems, Inc., Goleta, CA, USA). The data were full-wave rectified and averaged with a 100 ms time constant to draw the amplitude of the signals. For the sEMG signal processing, root-mean-square (RMS) operations were performed, and the value was calculated to validate the utility of the sEMG signals. The signal-to-noise (SNR) was then calculated using Eq. (1):

$$SNR = RMS_{signal} \Big/ RMS_{noise} \tag{1}$$

Before testing a muscle contraction, the baseline EMG signals were collected in the supine position for 10 s to measure the baseline electrode noise. A single-joint knee extension was conducted with the subject seated to induce muscle contraction signals. The subject extended the knee until the lower leg became parallel to the floor. Testing consisted of five consecutive trials for knee extension and knee flexion and each phase lasted for 5 seconds. In this study, individual differences in muscle activities were not considered because the study was performed on only one subject. The entire experiment was conducted at room temperature under 60% RH. The sEMG signals were recorded during the trials, and three contractions among five times, except the first and last trials, were used to calculate the average activated EMG for comparison (Kim et al., 2020, 2022a).

Results and Discussion

Morphology of the embroidery textile-based electrodes by stitch technique and distance

Table 3 lists the morphology according to the embroidery stitch technique and stitch distance. In this study, embroidery electrodes were fabricated in one layer to check the differences according to the stitching technique and density. As investigated by stitch type, the lock stitch has the same front and back sides because the lock stitch technique is a linear stitch formed by crossing the upper and lower yarns. On the other hand, in the case of a moss stitch, the front and back sides are different. The front side has a loop-shaped stitch and forms a bulky, three-dimensional structure. However, the back side is identified as a linear stitch in the same way as the lock stitch. The moss stitch technique forms a loop at the top while pulling a thread using a hook-type needle for one conductor and forms a stitch-type at the backside.

As confirmed by the stitch distance, the density increased as the distance between the conductive yarn narrowed. In particular, because the moss stitch has a bulky shape, the contact area between the conductive yarns increased as the stitch distance increased. This was also related to the amount of conductive yarn used, and as the density increased, the conductive yarn use of LS_1, LS_2, and LS_3 was 65.8 cm, 34.8 cm, and 22.0 cm, respectively. The corresponding use of MS_1, MS_2, and MS_3 moss was 263.2 cm, 139.2 cm, and 88.0 cm, respectively. The amount of yarn used in the stitch was approximately four times longer than the lock stitch. In addition, more conductive yarn was used in the case of a loop with a height of 3 mm higher than that of a linear stitch.

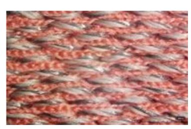
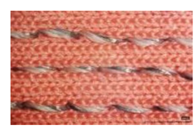
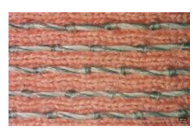
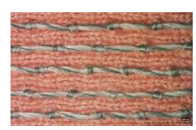
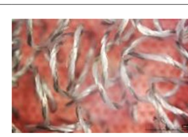
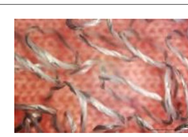
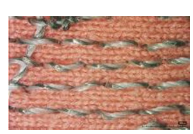
Figure 4 shows 3D images of LS_1 and MS_1 taken by X-ray microscopy. As shown in Fig. 4a, b, while LS_1 is linear, MS_1 has more conductive areas in the form of a loop. Moreover, the contact points between the conductive yarns increased. Figure 4c shows the height of the loop, showing that the overall height is approximately 3 mm. The morphological characteristics according to the stitching technique and density are expected to affect the tactile, sheet resistance of the electrodes, skin–electrode impedance, and sEMG signal acquisition.

KES-FB analysis of the embroidery textile-based electrodes by stitch technique and distance

Surface property

Embroidery is a patterning method by which a series of overlapping stitches is formed from a single continuous threador multiple threads, on a fabric substrate. As confirmed in the previous morphology result, it was possible to confirm the morphology of the

Table 3 Morphology of the embroidery-based textile electrodes by the stitch technique and distance

	Sample code		
	LS_1	**LS_2**	**LS_3**
× 35.0 (surface)			
× 35.0 (back)			
Stitch distance (mm)	1.0	2.0	3.0
*Total length (cm)	65.8	34.8	22.0
	Sample code		
	MS_1	**MS_2**	**MS_3**
× 35.0 (surface)			
× 35.0 (back)			
Stitch distance (mm)	1.0	2.0	3.0
*Total length (cm)	263.2	139.2	88.0

Total length*: (LS, Lock stitch) Number of stitch × (Stitch length + thickness of raw material)

(MS, Moss stitch) Number of stitch × 2(Loop height + thickness of raw material)

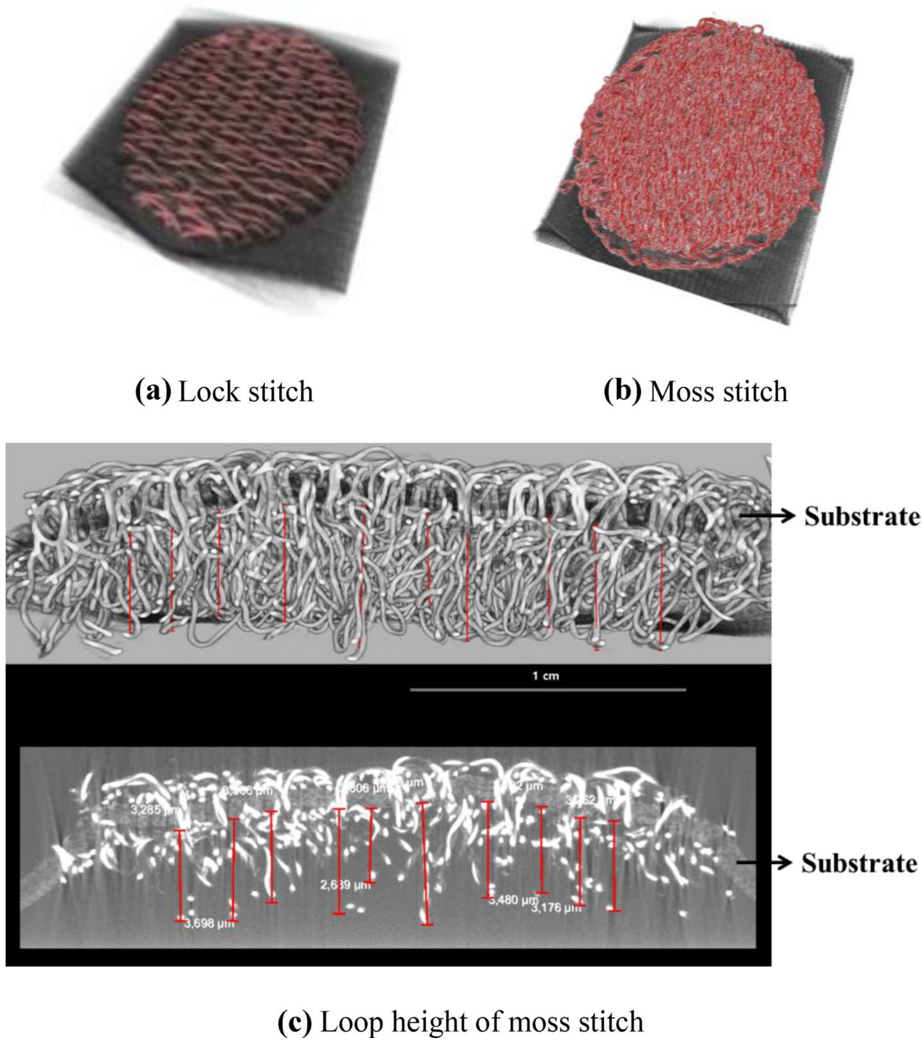


Fig. 4 3D image of (**a**) lock and (**b**) moss-based embroidery textile electrode and (**c**) loop height of the moss stitch

linear 2D and bulky 3D forms according to the stitch technique. Accordingly, Fig. 5 shows the measurement results of the surface properties according to the stitch technique and distance to quantify the morphological characteristics using the Kawabata system.

The surface characteristic values of the fabrics are as the coefficient of friction (MIU), mean deviation of MIU (MMD), and geometrical roughness (SMD), which are related to the smoothness of the fabric. The coefficient of friction is closely related to crispness, which is a feeling when the surface is crisp and rough, among the characteristics that evaluate the fabric quality and the coolness and warmth of the fabric. In addition, the smaller the MIU and SMD values, the smoother the surface is, and the larger the MIU and SMD values, the rougher the surface.

Figure 5a shows the results of MIU according to the stitch technique and distance in the wale and course directions. In the case of the substrate fabric and LS samples, the MIU values in the wale direction were small. On the other hand, the MS sample showed

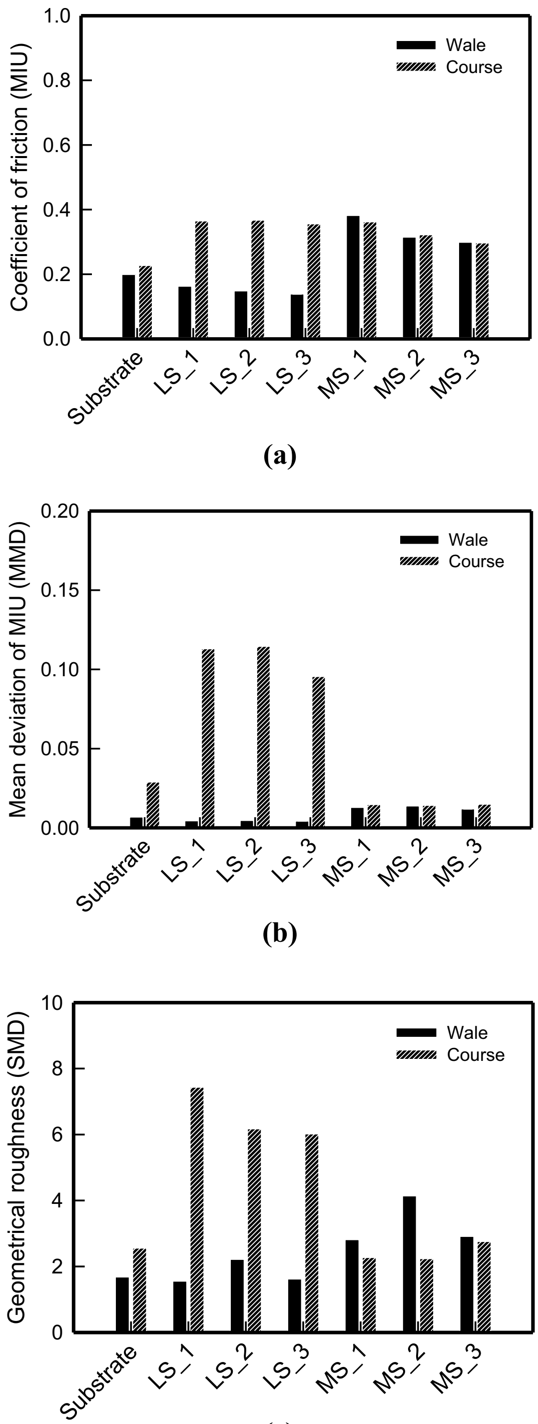
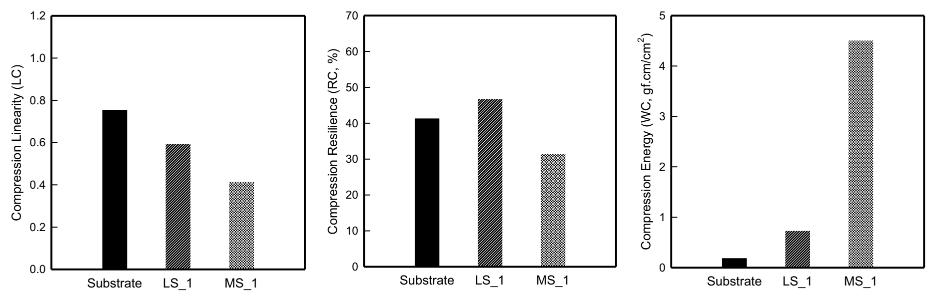


Fig. 5 Surface properties of the embroidery-based textile electrode using the stitch technique and distance.
* LS: Lock stitch, MS: Moss stitch

a similar MIU values in the wale and course directions. In LS, there is a difference between the substrate fabric and the conductive yarn in the electrode, so the difference in the wale and course directions appears large. However, the MIU values in the wale



(a) Compression linearity (b) Compression resilience (c) Compression energy

Fig. 6 Compression properties of the embroidery-based textile electrode using the stitch technique and distance. * LS: Lock stitch, MS: Moss stitch

Table 4 ΔT and WC/T0 value of the embroidery-based textile electrodes by the stitch technique and distance.

	Sample code		
	Substrate	**LS_1**	**MS_1**
ΔT*(%)	9.63 ± 0.92	26.55 ± 6.76	65.03 ± 4.24
WC/T0**	0.18 ± 0.03	0.37 ± 0.06	0.67 ± 0.08

*ΔT: (T0** – TM***) × 100

**T0: Thickness at 0.5 gf/cm^2

***TM: Thickness at maximum load

LS: Lock stitch, MS: Moss stitch

and course directions were similar in MS because the electrode is filled with conductive yarn. In case of MMD values by the embroidered technique, the deviation of LS tended to be larger than that of MS (Fig 5b).

In the SMD results, the lock stitch values were larger in the course direction, similar to the MIU results (Fig 5c). On the other hand, in the case of the moss stitch, the wale and course directions were relatively similar, and the course direction was relatively small. Accordingly, it was confirmed that the MIU of MS was similar to that of LS, but the difference according to the direction was small, and the deviation was also low.

Compressional property

Compressional property presents the resilience and fullness of the fabric, which are important parameters for determining comfort and hand value. In Kawabata's evaluation system, the factors that indicate compressional properties include compression linearity (LC), compressional energy (WC), which indicates the deformation with respect to compression force, and compression resilience (RC), which indicates the degree of recovery against compression deformation (Lee, 2022; Choi & Ashdown, 2000).

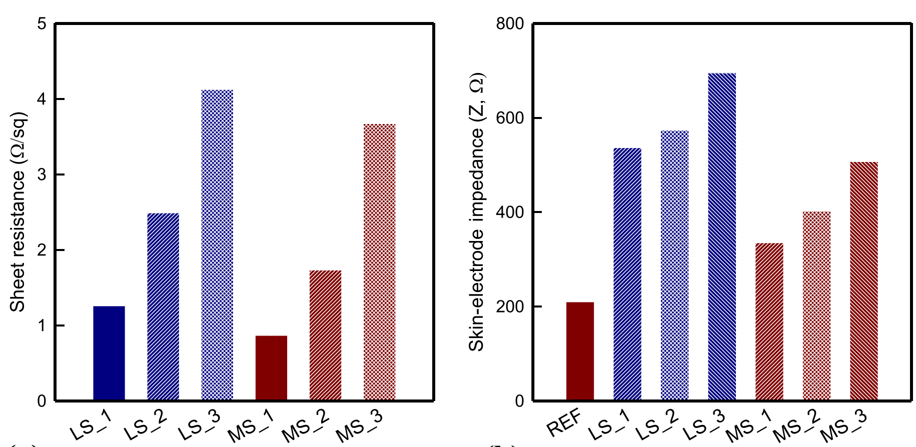
Figure 6 indicates the results of compressional properties of the embroidery-based textile electrode of substrate fabric, LS_1, and MS_1. A larger LC value means greater resistance to the compressive force. As shown in Fig. 6a, LC increases in the order of MS < LS < Substrate. MS_1 had the smallest LC value of 0.42 ± 0.08 compare with LS and

Substrate. Accordingly, the moss stitch has lower compression resistance than the lock stitch.

In general, the results of RC and WC mainly affect the characteristics of the fabric, and a larger RC value means greater recovery from compression. Compressional energy (WC) represents the ease of deformation with respect to the compressive force; this is related to the comfort of the textile product (Tadesse et al., 2018). The compressibility becomes easier with a lower compressional energy. As shown in Fig. 6b, RC appeared in the order of MS < LS < Substrate. The RC value of LS_1 had 46.8 ± 7.4%, and MS_1 showed 31.6 ± 2.1%. Figure 6c indicates the compression resilience of the embroidery textile electrode from the stitch technique and distance. WC was observed in the order of Substrate < LS < MS. Substrate showed the smallest value of 0.19 ± 0.03 gf cm/cm^2, and LS_1 showed approximately 0.76 gf cm/cm^2. The WC value of MS_1 was 4.51 ± 0.17 gf cm/cm^2. It was confirmed that the 3D loop-shaped MS consumes more compression energy than the 2D-shaped LS.

To relatively compare the stitch technique and density of embroidery electrodes, Table 4 lists the thickness change rate and WC by dividing the thickness (T0) of the sample. A comparison of the thickness before and after compression revealed Substate < LS < MS: 9.63 ± 0.92% for Substrate, 26.55 ± 6.76% for LS_1, and 65.03 ± 4.24% for MS_1. Therefore, it was confirmed that the thickness change was the largest during the compression of the moss stitch. As confirmed by the morphology, the moss stitch has a large range of heights that can be compressed under the same load because the moss stitch has a three-dimension structure with a loop shape. In contrast, the linear-type lock stitch is relatively compressed because it is close to the fabric. Thus, it was found to be a small range of LS. In addition, in the case of embroidery stitch, because it consists only of conductive yarn, it has a stiffer characteristic than general yarn. As the density increases in WF_MS, the amount of conductive yarn also increases, which appears to affect recovery (Lee, 2022).

As shown in Table 4, WC/T0 showed a similar trend to ΔT: Substrate < LS < MS. The WC/T0 values of LS_1 and MS_1 were 0.37 ± 0.06 and 0.67 ± 0.08, respectively. In other



(a) (b)

Fig. 7 (**a**) Sheet resistance and (**b**) skin–electrode impedance of an embroidery-based textile electrode using the stitch technique and distance. * LS: Lock stitch, MS: Moss stitch words, the embroidery-based textile electrode made of moss stitch was bulky, so it had great flexibility and a soft feel when compressed.

Sheet resistance of the embroidery textile-based electrodes by stitch technique and distance

The surface resistance is associated with the contact resistance between the neighboring yarns through contact points as crossovers (Katerine et al., 2022). Figure 7a shows the sheet resistances of the embroidery-based textile electrodes according to the stitch technique and distance. LS_1, LS_2, and LS_3 were 1.26 ± 0.16 Ω/sq, 2.49 ± 0.71 Ω/sq, and 4.12 ± 0.79 Ω/sq, respectively; MS_1, MS_2, and MS_3 were 0.87 ± 0.23 Ω/sq, 1.73 ± 0.47 Ω/sq, and 3.67 ± 0.89 Ω/sq, respectively. The sheet resistances of LS were larger than MS. In addition, as the stitch distance decreased, the resistance tended to decrease because conducting silver particles came close to the stitch distance, and more continuous conductive networks formed (Lee, 2022; Montazer & Komeily, 2015). Therefore, as confirmed in the morphology, there are many areas where the bulky moss stitch can contact each other compared to the linear lock stitch when compressed. The electrical properties are promoted by producing more networks that allow the movement of electrons. In addition, a higher yarn density means that the electrode area is more densely covered with conductive yarns. Therefore, the full electrode area is available as the contact area, while for a lower density, gaps are visible between the yarns, which means a difference between the electrode area and the effective contact area is expected. This suggests that a higher yarn density leads to better skin contact and improves electrode performance (Euler et al., 2021). Accordingly, the shape of MS_1 has the lowest skin–electrode impedance, enabling stable signal acquisition.

Skin–electrode impedance of the embroidery textile-based electrodes by stitch technique and distance

Because bio-potential signals exhibit a frequency range, contact impedance properties, which depend on signal frequency, must be characterized to provide a better understanding of the electrical properties at the skin–electrode interface. Accordingly, the skin–electrode contact impedance must be minimized to acquire a high-quality signal with low noise levels (Katerine et al., 2022).

Figure 7b indicates the skin–electrode impedance of embroidery-based textile electrodes using the stitch technique and distance. As shown in Fig. 7b, the skin–electrode impedance of the reference electrode, the Ag/AgCl hydrogel electrode, is 209.9 ± 4.5 Ω. The textile-type embroidery electrode showed less noise than the reference electrode. Textile electrodes generally show much higher impedance than conventional wet/gel electrodes when first applied to the skin owing to the strong capacitive behavior of dry textile electrodes from the absence of an electrolyte (Katerine et al., 2022). The lock and moss stitch-based embroidery electrode showed the same trend as the sheet resistance analyzed previously. Overall, the skin–electrode impedance of the moss stitch shows a lower value than the lock stitch sample, confirming that the noise is low. The LS_1, LS_2, and LS_3 skin–electrode impedances were 536.6 ± 11.8 Ω, 573.7 ± 18.8 Ω, and 694.9 ± 52.6 Ω, respectively; MS_1, MS_2, and MS_3 were also 335.1 ± 25.8 Ω, 402.3 ± 10.3 Ω, and 507.2 ± 24.5 Ω, respectively. Both results showed a gradual decrease

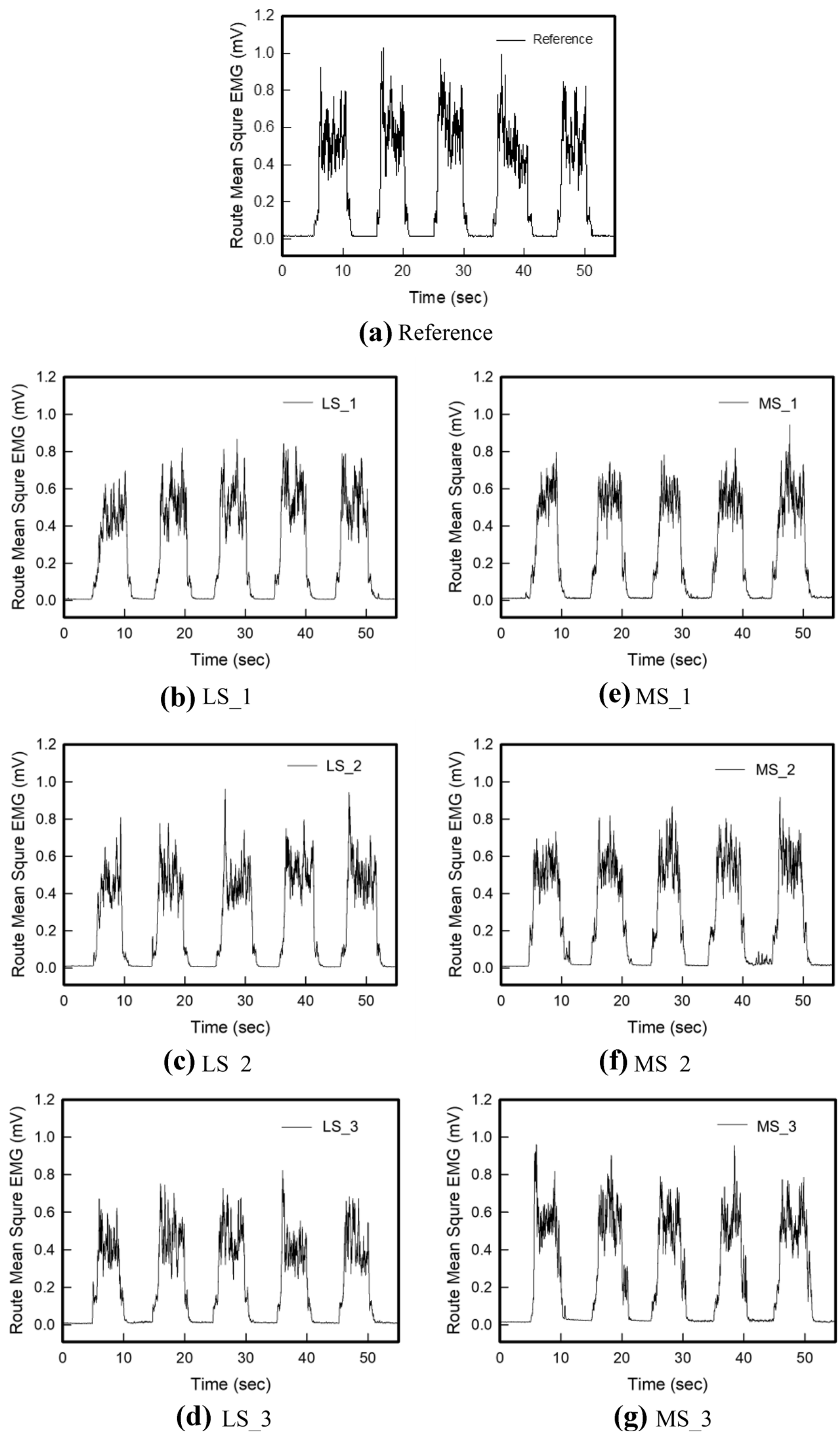


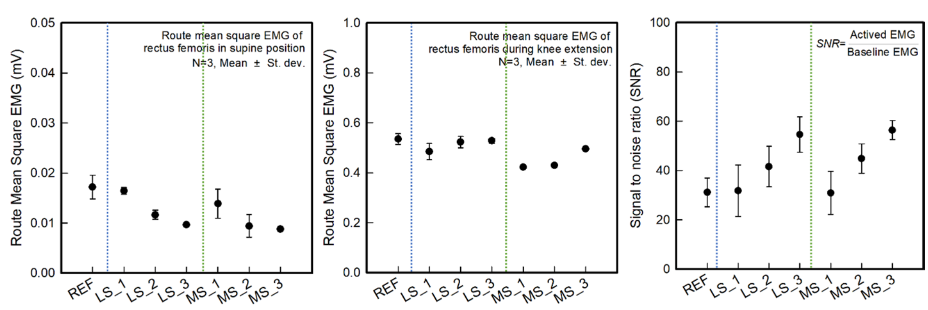
Fig. 8 Graph of filtered (20–500 Hz) in analog and full-wave of sEMG signals obtained via various electrodes: (**a**) reference, (**b**) LS_1, (**c**) LS_2, (**d**) LS_3, (**e**) MS_1, (**f**) MS_2, and (**g**) MS_3, respectively. * LS: Lock stitch, MS: Moss stitch

with decreasing the stitch distance. Accordingly, as the stitch distance decreased, the skin–electrode impedance decreased, confirming that a more stable signal could be obtained.

sEMG signal and average rectified sEMG of the embroidery textile-based electrodes by stitch technique and distance

Figure 8 shows the filtered from 20 to 500 Hz and full-wave sEMG signals obtained using different types of electrodes. According to previous studies (Kim et al., 2020, 2022a), the sEMG signal of rectus femoris was acquired by repeating the knee flexion–extension process five times while wearing a leg sleeve containing the embroidery textile electrodes. The sEMG signal was obtained from each electrode by repeatedly measuring for 5 seconds of release-and-rest periods following each flexion.

As shown in Fig. 8, all the embroidery electrodes acquired the sEMG signal of muscle activation. The signal amplitude of the reference electrode, which is the Ag/AgCl hydrogel electrode, was the largest, which was related to the impedance analyzed above. This is because textile electrodes generally show much higher impedance than conventional wet/gel electrodes when first applied to the skin, owing to the strong capacitive behavior of the dry textile electrodes from the absence of an electrolyte (Katerine et al., 2022). When comparing the stitch techniques of the embroidery-based textile electrode, the signal amplitude was similar to the lock and moss stitch, but the stability of the signal acquired, like noise generation, was more stable in the moss stitch. It also showed different results depending on the stitch distance. As the stitch distance became closer, the noise decreased during signal acquisition. Accordingly, the signal stability of LS_1 and MS_1 was excellent, and in particular, MS_1, which is a moss stitch, was the best. The sEMG signal shows a similar tendency to the skin–electrode impedance because it is affected by the skin adhesion of the electrode and the area where the conductive material in the electrode contacts the skin. Accordingly, from a structural point of view, the lock stitch has a 2-dimensional linear structure that allows excellent skin adhesion. On the other hand, when a leg sleeve with a 3-dimensional loop-type moss stitch is worn,



(a) Baseline electrode noise **(b)** sEMG amplitude during muscle contraction exerted by knee extension **(c)** Signal-to-noise ratio (SNR)

Fig. 9 Average rectified sEMG of the baseline during muscle activation for electrode type using different stitch techniques and distances: (**a**) baseline electrode noise, (**b**) sEMG amplitude during muscle contractions exerted by knee extension, (**c**) signal to noise (SNR). * LS: Lock stitch, MS: Moss stitch the contact points between the conductive yarns increase as they are closely adhered to by the clothing pressure. Thus, the signal was amplified and stable because more areas can be in close contact with the skin than the lock stitch.

Figure 9 presents the average rectified sEMG of the baseline before and after muscle activation and the signal-to-noise ratio (SNR) for the different electrode types. As shown in Fig. 9a, the baseline electrode noise differed according to the stitch technique and distance. The baseline noise of the Ag/AgCl electrode, which is the reference electrode, was 0.017 ± 0.004 mV. In the case of the embroidery textile electrodes, LS_1, LS_2, and LS_3 were 0.010 ± 0.000 mV, 0.012 ± 0.001 mV, and 0.016 ± 0.001 mV, respectively, and MS_1, MS_2, and MS_3 were 0.008 ± 0.000 mV, 0.009 ± 0.002 mV, and 0.014 ± 0.002 mV, respectively. As a result, the baseline electrode noise was lower than that of the Ag/AgCl electrode, suggesting that the embroidery-based textile electrodes were stable in the femoris in the supine position.

The results of the sEMG signals during muscle contraction indicated different values depending on the type of textile-base embroidery electrode (Fig. 9b). As shown in Fig. 9b, the reference electrode showed a value less than 0.535 ± 0.051 mV, and the embroidery textile electrode indicated a value less than that, but a slight difference was shown. LS_1, LS_2, and LS_3 were 0.527 ± 0.010, 0.523 ± 0.024 mV, and 0.485 ± 0.032 mV; MS_1, MS_2, and MS_3 were 0.495 ± 0.005 mV, 0.429 ± 0.006 mV, and 0.422 ± 0.005 mV, respectively. The results confirmed that the muscle activation signal amplitude tended to increase with decreasing stitch distance, and the moss stitch type showed a lower value than the lock stitch type. When comparing deviations, however, more stable signal acquirement was possible because the shape of the moss stitch is less than that of the lock stitch. As mentioned above, the 2D type of linear lock stitch has excellent skin adhesion, but it appears to generate more noise because the amount of conductive material in contact with the skin is relatively small compared to the moss stitch. In addition, When the loop-type moss stitch was worn as a leg sleeve, the loops were in contact flat with the skin while being pressed by the pressure of clothing, and the contact points between the conductors increased, so the area that could be in close contact with the skin is larger than the lock stitch. This suggests that the signal is stable and amplified when fabricated with a moss stitch.

Based on these results, the SNR was calculated as the ratio of active EMG to the baseline EMG. As shown in Fig. 9c, the SNR of reference electrode was 31.097 ± 4.328, LS_1, LS_2, and LS_3 were 54.609 ± 7.223, 41.616 ± 10.372, and 31.808 ± 8.245, respectively; MS_1, MS_2, and MS_3 were 56.354 ± 3.874, 44.815 ± 8.754, and 30.867 ± 5.982, respectively. The results of SNR confirmed that lock and moss stitches showed similar values for each stitch distance. In addition, in the case of standard deviation, the stitch distance was 1 mm < 2 mm < 3 mm, and the moss stitch showed a smaller value than the lock stitch. The conductive yarn in the electrode increased as the stitch distance decreased; thus, the area that could be in contact with the skin increased and stably adhered. Therefore, by applying a moss stitch when fabricating textile-type electrodes, it is expected to apply to the production of smart clothing for sEMG measurement, which is softer to the touch and more flexible and has excellent sEMG signal acquisition performance.

Conclusions

This study was proposed a dry electrode that can collect bio-signal of the human body using a traditional embroidery technique, a lock stitch, and a moss stitch that forms a three-dimensional structure while the yarn forms a loop. The performance according to the embroidery stitch technique and the stitch distance was evaluated by preparing samples with two stitch techniques, lock and moss stitch, and three stitch densities. The tactile, electrical, and EMG signal acquisition performance of each manufactured electrode was analyzed.

The surface and compressive properties of the electrodes were evaluated using the Kawabata evaluation system. As the stitch distance decreased, the density between the conductive yarns in the electrode increased. It was confirmed that the MIU was larger than the lock stitch because the moss stitch took the form of a loop. However, as the results of SMD analysis, the moss stitch showed a smaller value than the lock stitch, and it was confirmed that the surface of MS was smoother than that of LS. In the case of compression property, the height range that can be compressed under the same load was large because the moss stitch is a three-dimension type having a loop shape in structure. On the other hand, the linear-type lock stitch has a relatively small compressive range because it is close to the fabric. Accordingly, the WC value of MS was small, confirming more flexibility. The shape and density of these electrodes also affected the sheet resistance and skin–electrode impedance. The stitch distance decreased, and the sheet resistance and skin–electrode impedance of the electrode fabricated with the loop-type moss stitch decreased. This was attributed to more paths for silver particles with conductivity to form a conductive network. In addition, the area that can contact each other increases in the bulky form than in the linear form. Accordingly, it affected sEMG signal acquisition, and it was confirmed that more stable signal collection was possible in the case of moss stitches.

Therefore, by applying moss stitch when fabricating textile-type electrodes, it is expected to apply to the production of smart clothing for sEMG measurement, which is softer to the touch and more flexible and has excellent sEMG signal acquisition performance.

Authors' contributions

DL conceived the work and SR and HK prepared the samples and SR and HK performed the experiments. HK and WJ are participated in the sequence alignment and drafted the manuscript. All authors read and approved the final manuscript.

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Availability of data and materials

The data sets used and analyzed during the current study are available from the corresponding author on reasonable request.

Declarations

Ethics approval and consent to participate

This study was conducted under the approval of the Institutional Review Board of Korea Institute of Industrial Technology (KITECH) (IRB approval No. 2022-002-001) regarding ethical issues including consent to participate.

Competing interests

The authors declare that they have no competing interests.

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References

Ahsan, M., Teay, S. H., Sayem, A. S. M., & Albarbar, A. (2022). Smart clothing framework for health monitoring applications. *Signals, 3*, 113–145. https://doi.org/10.3390/signals3010009

Choi, J., & Hong, K. (2019). Compression suits with and without films and their effects on EMG during isokinetic exercise. *Fashion and Textiles, 6*, Article 20. https://doi.org/10.1186/s40691-019-0176-2

Choudhry, N. A., Arnold, L., Rasheed, A., Khan, I. A., & Wang, L. (2021). Textronics—a review of textile-based wearable electronics. *Advanced Engineering Materials, 23*(12), Article 2100469. https://doi.org/10.1002/adem.202100469

E-textile based modular sEMG suit for large area level of effort analysis

Euler, L., Guo, L., & Persson, N.-K. (2021). Textile electrodes: Influence of knitting construction and pressure on the contact impedance. *Sensors, 21*, Article 1578. https://doi.org/10.3390/s21051578

Gizem, A., Ozberk, O., Ata, J. G., Tamador, A. E., Karl, B., & Murat, K. Y. (2019). Wearable and flexible textile electrodes for biopotential signal monitoring: A review. *Electronics, 8*(5), Article 479. https://doi.org/10.3390/electronics8050479

Goncy-Berk, G., & Tuna, B. G. (2021). The effect of sleeve pattern and fit on e-textile electromyography (EMG) electrode performance in smart clothing design. *Sensors, 21*, Article 5621. https://doi.org/10.3390/s21165621

Katherine, L., Harishkumar, N., Amir, S., Addie, B., Saeid, S., Peynan, S., & Frank, K. (2022). Electronic textiles for electrocardiogram monitoring: A review on the structure-property and performance evaluation from fiber to fabric. *Textiles Research Journal 93*(3-4), 878–910. https://doi.org/10.1177/00405175221108208

Kim, H., Kim, S., Lim, D., & Jeong, W. (2022a). Development and characterization of embroidery—based textile electrodes for surface EMG detection. *Sensors, 22*, Article 4746. https://doi.org/10.3390/s22134746

Kim, H., Rho, S., Han, S., Lim, D., & Jeong, W. (2022b). Fabrication of textile–based dry electrode and analysis of its surface EMG signal for applying smart wear. *Polymers, 14*, Article 3641. https://doi.org/10.3390/polym14173641

Kim, S., Lee, S., & Jeong, W. (2020). EMG measurement with textile-based electrodes in different electrode sizes and clothing pressures for smart clothing design optimization. *Polymers, 12*, Article 2406. https://doi.org/10.3390/polym12102406

Lam, E., Alizadeh-Meghrazi, M., Schlums, A., Eskandarian, L., Mahnam, A., Moineau, B., & Popovic, M. R. (2022). *Journal of Rehabilitation and Assistive Technology Engineering, 9*, 1–18. https://doi.org/10.1177/20556683211061995

Lee, S. H. (2022). Analysis of electrical and comfort properties of conductive knitted fabrics based on blending ratio of silver-coated yarns for smart clothing. *Journal of Engineering Fibers and Fabrics, 17*, 1–14. https://doi.org/10.1177/15589250221104474

Mecnika, V., Hoerr, M., Krievins, I., Jockenhoevel, S., & Gries, T. (2014). Technical embroidery for smart textiles: Review. *Materials Science Textiles and Clothing Technology, 9*, 56–63. https://doi.org/10.7250/mstct.2014.009

Montazer, M., & Koneily Nia, M. (2015). Conductive nylon fabric through in situ synthesis of nano-silver: Preparation and characterization. *Materials Science Engineering C, 56*, 341–347. https://doi.org/10.1016/j.msec.2015.06.044

Rho, S. H., Lee, S., Jeong, W., & Lim, D.-Y. (2022). Study of the optimization of embroidery design parameters for the technical embroidery machine: Derivation of the correlation between thread consumption and electrical resistance. *Textile Research Journal, 92*(9–10), 1550–1564. https://doi.org/10.1177/00405175211061028

Shuvo, I. I., Shah, A., & Dagdeviren, C. (2022). Electronic textile sensors for decoding vital body signal state-of-the-art review on characterizations and recommendations. *Advanced Intelligent Systems, 4*, Article 2100223. https://doi.org/10.1002/aisy.202100223

Tadesse M G, Chen Y, Wang L, Nierstrasz V, Loghin C. (2018). Tactile comfort evaluation of conductive knitted fabric using KES-FB. IOP Conference Series Materials Science Engineering. 374, pp.012056. https://doi.org/10.1088/1757-899X/374/1/012056

Water-retentive, 3D knitted textile electrode for long-term and motion state bioelectrical signal acquisition

Zahid, M., Rathore, H. A., Tayyab, H., Rehan, Z. A., Rashid, I. A., Lodhi, M., Zubair, U., & Shahid, I. (2022). Recent developments in textile based polymeric smart sensor for human health monitoring: A review. *Arabian Journal of Chemistry, 15*, Article 103480. https://doi.org/10.1016/j.arabijc.2021.103480

Zheng, Z., Wu, Z., Zhao, R., Ni, Y., Jing, X., & Gao, S. (2022). A review of EMG-, FMG-, and EIT-based biosensors and relevant human–machine interactivities and biomedical applications. *Biosensors, 12*, Article 516. https://doi.org/10.3390/bios12070516

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