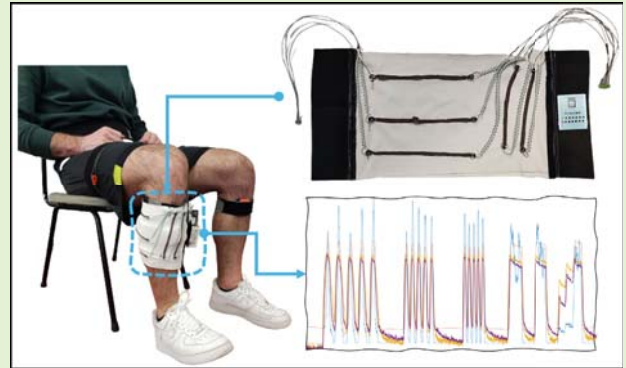


MuscLab: A Flexible and Elastic e-Textile Band for Monitoring Muscle Contraction

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Abstract—Mecanomyographic (MMG) and force myography (FMG) sensors emerged as an alternative to electromyography (EMG) for monitoring muscle contraction without requiring the direct contact of the sensor with the user's skin. Nonetheless, most of the available sensors 1) require as many sensors as there are muscles to monitor, resulting in a time-consuming preparation for practical use and 2) present a nonelastic nature, entailing a customized design to the user's segment anthropometries. Thereby, we developed the MuscLab system, which uses e-textile (piezoresistive textile) sensors sewn onto a flexible and elastic textile band. It simultaneously monitors and discriminates muscle contractions across different muscle groups in individuals with a shank perimeter ranging from 33.5 to 48.7 cm. From a benchmark analysis involving ten non-disabled individuals, the MuscLab was able to 1) detect muscle contractions at different motion cadences, with an average delay of 135.8 ± 78.0 ms and strong spearman correlation (0.78 ± 0.08) regarding EMG signals of tibialis anterior (TA) and gastrocnemius lateralis (GALs) and 2) distinguish different levels of muscle contraction performed at different ankle joint angles (the average coefficient of determination of 0.92 ± 0.10 and 0.89 ± 0.16 for ankle dorsiflexion and plantar flexion movements, respectively). This research advances with new insights regarding the use of e-textiles in an elastic band for monitoring the multimuscle contraction of users with different anthropometries.

Index Terms—Electromyography (EMG), e-textiles, muscle contraction, piezoresistive sensor.



I. INTRODUCTION

ACCORDING to the World Health Organization, musculoskeletal disorders (MSDs) are the world's leading

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cause of human motor disability. The estimated 1.71 billion people (one in four people) live with MSDs [1]. Most MSDs are associated with the work context, sedentary lifestyle, and practice of sports, at a professional or amateur level [1]. This condition makes it difficult to perform daily household and work tasks and it corresponds to the main reason for absenteeism and early retirement [2]. The high prevalence of MSDs raises the need for developing solutions to promote objective, noninvasive, and real-time monitoring of muscle contraction and relaxation. Ideally, this solution should be versatile to use in different contexts, such as 1) health, to support the clinical diagnosis of the evolution of muscle contraction; 2) sports, for the analysis of muscle performance and prediction of the risk of MSDs; and 3) work, to support the ergonomics assessment of muscle condition and prediction of the risk of MSDs.

Typically, surface muscle contraction monitoring is conducted by expensive but high-precision sensing electromyography (EMG) equipment. The gold-standard surface EMG solutions include the noninvasive EMG electrodes from Noraxon [3] and Cometa [4] or the dry EMG sensors from Delsys [5]. However, the long-term use of these sensors can be affected by 1) sweating; 2) temperature variations; and 3) movements between the skin and the electrodes [6]. Furthermore, if several agonist and antagonist muscles need

to be monitored simultaneously, it will need as many sensors as the number of muscles to monitor. This may result in a more time-consuming solution for donning and doffing, narrowing its practical use.

As an alternative to overcome the limitations of using EMG, mechanomyography (MMG) and force myography (FMG) have emerged for muscular analysis [10], [11], [12], [13], [14], [15], [16], [17], [18], [19], [20]. MMG sensors do not measure the electrical activity of muscles but instead detect mechanical vibrations associated with muscle contractions. These sensors can identify whether a muscle is contracting or relaxing without requiring direct contact with the user's skin. The principle underlying MMG is that muscle contraction typically results in muscle shortening, which generates mechanical vibrations. Accordingly, MMG sensors (such as inertial sensors, piezoelectric sensors, or microphones [10], [11]) positioned above the muscle can detect these vibrations and monitor muscle contraction. In contrast, FMG sensors monitor muscle contractions by measuring the changes in muscle cross-sectional area, stiffness, or tension. This is achieved using sensors, such as force-sensitive resistors [12], [17], [18], piezoresistive materials [15], [16], triboelectric nanogenerator (TENG) [13], strain gauges [20], capacitive [14], or optical sensors [19], which capture the volumetric changes caused by muscle bulging during contraction.

While studies [10], [11], [12], [13], [14], [18], [19], [20] offer valuable methods for monitoring muscle contractions in specific muscles, they are less practical for daily use when simultaneous monitoring of multiple muscle groups is required, as the number of sensors increases proportionally to the muscles to monitor.

Advancing these approaches, studies [15], [16], [17] introduced innovative solutions for monitoring the contraction of multiple muscles or muscle groups simultaneously. The study [15] proposed a cord-shaped piezoresistive sensor placed around the forearm to detect muscle contractions. While effective for monitoring agonist and antagonist muscles together, this approach cannot distinguish between their individual contractions, as both cause the cord to stretch.

On the other hand, the study [16] disclosed a flexible textile band (72% nylon and 28% spandex) for monitoring the muscle activity of the muscles in the shank segment, while discriminating which muscle is contracted. A flexible, piezoresistive, and conductive e-textile has been integrated into an external textile band and sewn into a matrix shape. The external textile band was designed for the shank segment and, due to its matrix configuration, it is possible to discriminate which muscles of the shank region are contracted/relaxed through the pressure they make on the different zones of the matrix. Similarly, in the patent document [17], a circular band with pressure sensors is described for the simultaneous monitoring of muscle contraction/relaxation of muscles responsible for hand, wrist, or arm movements. Although these two approaches [16], [17] can monitor the muscle activity of several muscles simultaneously, the textile bands lack elasticity, necessitating user-specific devices to ensure proper sensor placement according to individual anthropometry. To the best of our knowledge, there is no flexible and elastic textile band with MMG or FMG

TABLE I
MUSCLAB REQUIREMENTS

Requirement	Value
Real-time data acquisition	100 Hz
Wireless data transmission	Bluetooth > 4.0
Low packet loss	< 5%
Synchronization with third-party systems	-
Rechargeable battery	Autonomy: > 2h
Built-in inertial sensor	Accelerometer ($\pm 16G$ full scale) and gyroscope (± 2000 °/s)
The electric circuit must be isolated from the user	Inner band
The electric circuit must be protected against possible external perturbations	-
The electric circuit must be coupled inside a small and compact case	Length: < 5.5 cm Width: < 4.5 cm Height: < 2.5 cm
Low mass	< 100 g
The textile band must be elastic and adjustable to different anthropometries	35 cm < Shank perimeter < 45 cm
Easy preparation	Donning and Doffing: < 2 minutes
The system must not limit the user's movement	-

sensing features. Advancing over the current literature [10], [11], [12], [13], [14], [15], [16], [17], [18], [19], [20], this study proposes the MuscLab system—a stand-alone FMG solution based on e-textile (piezoresistive textile) sensors sewn onto a flexible and elastic textile band. The MuscLab system was designed as a proof of concept to simultaneously monitor and discriminate the muscle contraction of the shank segment muscles [mainly, the tibialis anterior (TA) and gastrocnemius lateralis (GALs)] in individuals with different anthropometries. The system was validated against the EMG Delsys system during isotonic muscle contractions, i.e., dorsiflexion and plantar flexion movements of the ankle joint, performed by individuals with variable anthropometric features.

II. METHODOLOGY

A. System Requirements

The MuscLab system was designed through a user-centered approach, including features that favor its acceptance. To this end, the system should 1) be a stand-alone monitoring solution, incorporating the sensing, processing, transmitting, and power supply units into a single, compact, and noninvasive device and 2) use a wireless protocol to transmit data to external units, avoiding wired communication means to be easy-to-use. Additionally, the system must be 1) flexible, to accommodate different human movements without break and 2) elastic, presenting the ability to stretch and to return to its original shape as a response to muscle contraction and relaxation, respectively. Table I presents the functional and nonfunctional requirements set for developing the MuscLab system.

B. System Overview

MuscLAB consists of a flexible and elastic textile band (47% recycled polyamide, 25% nonrecycled polyamide, and

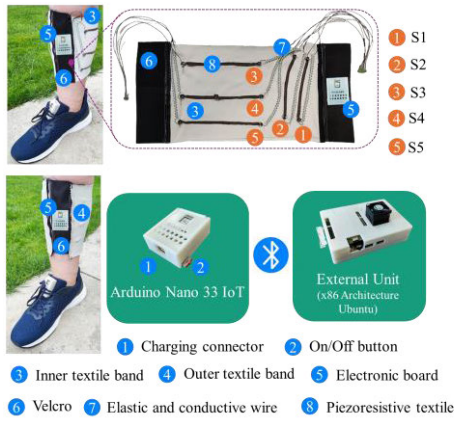


Fig. 1. MuscLab design. S1–S5 represent each one of the five piezoresistive textile strips, respectively.

28% elastane) with a double layer: an inner layer and an outer layer, as depicted in Fig. 1. The inner layer (Fig. 1-3) is instrumented with piezoresistive textile strips (Fig. 1-8) to assess muscle contraction and it ensures that the piezoresistive textile sensors do not contact with the user's skin. The outer layer (Fig. 1-4) guarantees the sensors' isolation from the environment. Both inner and outer layers are elasticated to facilitate adaptation to different anthropometries. These bands integrate a Velcro fastener to facilitate donning and doffing by the user (Fig. 1-6).

Additionally, the electronic board (Fig. 1-5) for powering the entire system and for acquiring, processing, and transmitting the sensor data to an external unit is coupled in a 3-D printed case that is placed over the Velcro area. The system also presents an inertial sensor that is embedded in the electronic board (described in Section II-C).

The external unit runs the MuscLab Application, which is responsible for receiving data from the MuscLab system via the Bluetooth communication protocol at 100 Hz.

C. Hardware

At the hardware level, MuscLab comprises three physical units: 1) control unit; 2) sensor unit; and 3) power supply unit, as depicted in Fig. 2.

1) **Control Unit:** The control unit includes 1) an Arduino Nano 33 IoT positioned in the electronic board (Fig. 1-5) and 2) an external unit, positioned in the same environment as a data collection is conducted. We used the wireless communication protocol with 4.2 technology (Bluetooth 4.2) to establish the communication between the Arduino Nano 33 IoT and the external unit. With respect to the Arduino Nano 33 IoT, it comprehends two individually programmed processors: 1) Cortex-M0 32-bit SAMD21, responsible for acquiring data from the sensors (inertial and piezoresistive textile sensors), calibrating the sensors, monitoring the battery level, and generating packets to send the sensor data to the external unit and 2) u-blox NINA-W102, responsible for handling the Bluetooth 4.2 communication with the external unit.

The external unit corresponds to a platform running the Ubuntu operating system. It must contain a processor and random access memory of at least 2.24 GHz and 4 GB,

respectively. In this case, the UDOO x86 was employed. The external unit is responsible for 1) receiving data from the inertial sensor, the piezoresistive textile, and the system's battery level; 2) filtering the piezoresistive textile sensor data; 3) storing all data in physical memory; and 4) quantifying muscle contraction.

2) **Sensor Unit:** The sensor unit includes an inertial sensor (integrated into the control board—Arduino Nano 33 IoT) and the e-textile, a configurable sensor made up of piezoresistive textile strips.

The inertial sensor (LSM6DSL) contains an accelerometer ($\pm 16G$ full scale) and a gyroscope (± 2000 °/s). It monitors 3-D angular velocity and linear acceleration of the limb on which it is placed. Since this sensor is already integrated into the Arduino Nano 33 IoT, it sends data through the I²C protocol.

The piezoresistive textile sensor consists of a flexible, elastic, and conductive fabric (i.e., e-textile sensor) with piezoresistive properties made of 78% polyamide and 22% elastane with a 99% pure silver bath: Shieldex¹ Technik-tex P130+B. We selected this piezoresistive textile sensor given its proper performance from a comparison of 16 piezoresistive textiles presented in the study [21]. It showed 1) no signs of fatigue during stretching; 2) the quickest responses between relaxing and stretching movements; 3) better elasticity and stability when relaxed; 4) the best linearity; and 5) low hysteresis ($<17\%$) [21]. Nonetheless, it is important to note that, according to the study [21], the chosen piezoresistive textile showed the signs of relaxation fatigue after 75 relaxation cycles. More details on the tests carried out on the selected piezoresistive textile can be found in [21].

As a proof of concept to monitor the muscular contraction of the shank segment muscles (mainly, the TA and GAL muscles), we developed a specific configuration composed of five piezoresistive textile strips (S1–S5, as depicted in Fig. 3). Two strips were arranged diagonally at the shank front (S1 and S2) and three strips were arranged horizontally at the shank back (S3–S5), all placed transversely to the direction of expansion of the mentioned muscles.

This configuration makes it possible to 1) determine the location of the muscle group that is contracted/relaxed (flexor or extensor groups); 2) determine when the contraction/relaxation occurs; and 3) quantify the contraction exerted by each muscle group.

Considering that the Shieldex Technik-tex P130+B has a small working range (from 56.4 to 62.7 Ω [21]), we designed an electronic setup optimized for the detection of small resistance variations and the linearization of the measured signals. For this purpose, piezoresistive textile strips were integrated into a Wheatstone bridge configuration. A digital potentiometer (AD8400—1 k Ω) was used to balance the Wheatstone bridge for each strip. The use of the digital potentiometer was essential to enable a fast and precise resistance adjustment as the strip's resistance varied slightly across strips and individuals. By allowing precise adjustments, the digital potentiometer ensured that the Wheatstone bridge

¹Registered trademark.

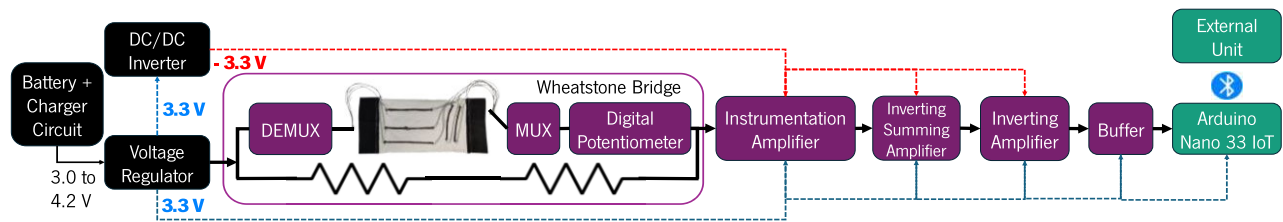


Fig. 2. MuscLab hardware diagram. The black, purple, and green boxes represent the power supply, sensor, and control units, respectively. DEMUX and MUX represent the demultiplexer and multiplexer, respectively.

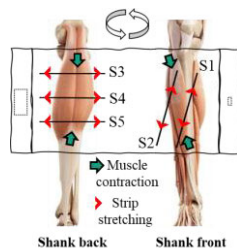


Fig. 3. Arrangement of S3–S5 in the shank back and S1 and S2 in the shank front.

was properly balanced for each strip, thereby improving measurement accuracy. In addition, to minimize the dimensionality of the electronics, a single Wheatstone bridge was shared between several piezoresistive strips. This was achieved using an eight-channel multiplexer/demultiplexer (74HC4051), allowing the system to interact sequentially with each strip, while reducing the number of electronic components and maintaining the compactness of the system, as shown in Fig. 2. The resistance variation in each strip was detected by an instrumentation amplifier (INA818) with a unitary gain. Then, an inverting summing amplifier was employed to add gain to the signal and an empirical offset of 0.75 V. The offset summing was done to ensure that there were no negative values before reading the signal through the 12-bit analog-to-digital converter (ADC) of the Arduino Nano 33 IoT. As the different strips can capture different levels of muscle contraction, different gain settings were required for each strip. To this end, the inverting summing amplifier incorporates another digital potentiometer (AD8400 50 k Ω). This enables the automatic adjustment of the gain per strip described in Section II-D. Then, an inverting amplifier with a gain of 2 was employed to obtain positive voltage values. At last, a rail-to-rail buffer powered between 0.0 and 3.3 V was used to prevent the ADC from damages related to possible negative voltages or voltages higher than the admissible value by the ADC, i.e., 3.3 V. All these surface mount device components were soldered onto a printed circuit board, together with the Arduino Nano 33 IoT and were coupled inside a 3-D printed case, depicted in Fig. 1-5.

The communication between the electronics and the piezoresistive textile strips takes place via a flexible, elastic, and conductive thread (Condustretch 5 ISO Ci-0.48 P2) composed of wires organized in a zigzag pattern (Fig. 1-7). By using a zigzag pattern, it is possible to maintain the desired elasticity, with a stretch capacity of 150%. This conductive

thread was chosen for three purposes. First, it presents a resistance equal to or less than 2.5 Ω /m, a value that remains invariable when the thread is stretched. Second, it is an insulated conductor wire, protecting it against possible external interference. Third, the elastic properties of the conductive wires meet elasticity requirements.

3) Power Supply Unit: The power supply unit includes a 4.2-V, 450-mAh rechargeable LiPo battery (voltage ranging from 4.2 to 3.1 V) connected to a battery charger (BQ24091DGQR) via a micro-USB connector (Fig. 1-1). This unit includes a voltage regulator and a dc/dc inverter to generate 3.3 and -3.3 V, respectively. While the 3.3-V voltage is used to power the Arduino Nano 33 IoT and as the positive voltage supply for all the integrated circuits, the -3.3 -V voltage is used as the negative voltage supply for the integrated circuits. Additionally, this unit presents an ON/OFF button to turn ON/OFF the system (Fig. 1-2).

D. Software

At the software level, two algorithms programmed in C++ code were developed: an Arduino code implemented on the Arduino Nano 33 IoT and a MuscLab application programming interface (API) implemented on the external unit. These two algorithms were programmed to communicate with each other through Bluetooth 4.2.

After switching on, the system performs a first calibration, adjusting the resistance of the digital potentiometer (AD8400 1 k Ω) on the Wheatstone bridge automatically. This calibration aims to balance the Wheatstone bridge. Since we have implemented an offset voltage of 0.75 V in hardware, the Wheatstone bridge is balanced when the voltage read by the ADC of the Arduino Nano 33 IoT is close to this offset voltage (0.75 V). To do this, we have implemented a software routine that adjusts the resistance of the digital potentiometer according to the resistance of each piezoresistive textile strip. During this calibration step, the user must remain in a sitting position and static for 10 s, while the software routine finds the optimal position among 256 possible programmable positions of the digital potentiometer until the output of the ADC reaches 0.75 V for each piezoresistive textile strip.

Once calibrated, the algorithm moves on to a calibration routine for the second digital potentiometer (AD8400 50 k Ω). This potentiometer is responsible for amplifying the signal read for each strip, allowing gains between 1 and 250. To calibrate it, the user was asked to perform dynamic contractions for 10 s. The purpose of this calibration was to determine the gain necessary to ensure that the maximum voltage is

3.3 V, the maximum value allowed by the Arduino. Moreover, through this calibration routine, we aim to guarantee that the prestressing of the band (while worn) does not affect the measurements.

After calibrating the digital potentiometers, the Arduino's program waits to receive a command message from the MuscLab API. These commands are received by the NINA processor and sent via universal asynchronous receiver/transmitter (UART) serial communication to the Cortex-M0 32-bit SAMD21 of the Arduino Nano 33 IoT. If the command sent by the API is a start command, it will initiate the collection of inertial data, piezoresistive textile data, and battery voltage level. This is done by reading the battery data every 5 min and the inertial and piezoresistive data every 10 ms. The data are sent to the API at 100 Hz through a data message, organized as follows: package ID, battery level, time stamp, three-axis accelerometer and gyroscope, and piezoresistive textile data. The piezoresistive data are filtered in the API by using a low-pass exponential filter with a cutoff frequency of 5 Hz. The API then saves all data into a text file. The code is continuously running until the user stops it through a stop command sent by the API.

E. Experimental Validation

We carried out two experimental tests to infer the operability of the MuscLab system. First, we conducted bench tests to check if the requirements defined for MuscLab were met. Second, we performed human tests to evaluate the reliability of the MuscLab signals against EMG and kinematic signals during dorsiflexion and plantar flexion movements.

1) Bench Tests: We carried out a benching protocol to evaluate the MuscLab system requirements, stated in Table I, regarding the 1) acquisition frequency (100 Hz); 2) percentage of packet loss of wireless communication (below 5%); and 3) autonomy (>2 h). The MuscLab mass and dimensions were also measured. The protocol consisted of two trials, where the MuscLab system acquired and transmitted the inertial and piezoresistive textile data at 100 Hz. The protocol started with a battery level of 100%. The battery level was measured with a multimeter at the beginning of the protocol and in 20-min intervals until the system switched off due to lack of battery power. During the data collection, the time stamp of each received sample at the API was registered. The trial duration was recorded by a chronometer to compute the battery autonomy.

After the data collection, the time between consecutive samples was computed. To infer the acquisition frequency, the average time between consecutive samples was determined. Furthermore, all samples received with a time between consecutive samples of less than 9.9 ms or more than 10.1 ms were considered lost samples. The percentage of lost packets was calculated taking into account the received and lost samples.

2) Human Tests: Ten healthy participants (five males and five females) were involved in the validation of the MuscLab (average age of 26.5 ± 2.8 years old, an average body mass of 65.4 ± 11.2 kg, and an average body height of 169.8 ± 10.9 cm). The lowest and highest shank perimeters measured were 35.0 and 42.1 cm, respectively, achieving an



Fig. 4. Participant instrumented with the MuscLab, Delsys, and Xsens systems.

average value of 37.6 ± 1.9 cm. Prior to the experiments, all participants gave written informed consent in accordance with the ethical guidelines established by the Ethics Committee of the University of Minho (CEICVS 006/2020).

We started the protocol by equipping the participants, as illustrated in Fig. 4. First, the participants were instrumented with the Delsys Trigno² Avanti system (Delsys, Massachusetts, USA) to capture EMG signals from two lower limb muscles: the TA and the GAL of the right leg. The sensors were positioned above the referred muscles following SENIAM recommendations [22]. Furthermore, the system was configured to compute the rms of the EMG signals at 100 Hz. Moreover, we equipped the participants with seven IMUs (torso, thighs, shanks, and feet) from the MVN BIOMECH system (Movella—Henderson, NV 89 052) to collect lower limb joint angles at 100 Hz.

Then, the participants were instrumented with the MuscLab system at the right shank. The MuscLab system was positioned above the sensors from Delsys and MVN systems, as depicted in Fig. 4. Participants were then asked to perform the N-pose +Walk calibration procedure from the MVN BIOMECH system. Participants stood in the N-pose for 3 s and then walked 5 m forward and backward in a straight line. Then, participants were asked to sit to perform the MuscLab calibration: remain static for 10 s and then perform repetitive dorsiflexion and plantar flexion movements for 5 s. Once calibrated, participants completed three independent tests per each movement type: dorsiflexion and plantar flexion movements.

The first test consisted of the following sequence: 1) started in a relaxed position for 5 s [Fig. 5(a)] and 2) carried out five dorsiflexion repetitions [Fig. 5(b)] at three cadences imposed by a metronome, with each cadence separated by a 5-s relaxed position. The chosen cadences were 40, 75, and 105 beats/min, which may correspond to the walking speeds of 1.0, 2.5, and 4.0 km/h [23]. This first test aimed to analyze the system's responsiveness to different motion cadences. For that, we computed the delay between the MuscLab signals and the signals collected by the Delsys and MVN systems for each cadence.

The second test included two constant dorsiflexion movements for 3 s each, separated by a 3-s relaxed position. This test aims to evaluate the MuscLab's measuring repeatability. In this test, for both the MuscLab and Delsys systems, we

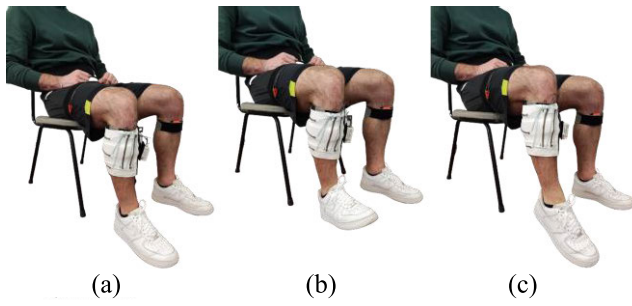


Fig. 5. Participant performing (a) relaxed movement, (b) dorsiflexion movement, and (c) plantar flexion movement.

computed the percentage of the average magnitude variation from the first to second muscle activations.

In the third test, participants were instructed to execute three constant dorsiflexion movements for 2 s each, at different ankle dorsiflexion angles, i.e., 1) a small dorsiflexion movement; 2) a dorsiflexion movement at an intermediate position between the relaxed and the maximum positions; and 3) a maximum dorsiflexion movement. This magnitude test was performed to verify if the system can discriminate different muscle contraction levels at different ankle joint angles. In this connection, we computed the coefficient of determination (R^2) between the MuscLab and the MVN signals. In all tests, the time was controlled by an external researcher.

Subsequently, the participants were asked to perform the same three tests but for plantar flexion movements [Fig. 5(c)].

In addition, we evaluated the correlation of the MuscLab system under the described dorsiflexion and plantar flexion movements with the EMG signals of the TA and GAL muscles, respectively. For that, we used the spearman correlation coefficient (r) to evaluate the strength and direction of the monotonic relationship between MuscLab and EMG signals. A p -value of 0.05 was used to determine whether the observed correlation between the two systems was statistically significant.

According to the study [24], the relationship can be categorized as 1) negligible if $0.0 < r < 0.19$; 2) weak if $0.20 < r < 0.29$; 3) moderate if $0.30 < r < 0.39$; 4) strong if $0.40 < r < 0.69$; and 5) very strong if $r \geq 0.7$.

III. RESULTS

A. Bench Tests

The MuscLab system showed an average 1) acquisition frequency of 100.0 ± 0.007 Hz ($\cong$ requirement of 100 Hz); 2) percentage of packet loss of $0.0972\% \pm 0.0318\%$ ($< 5\%$); and 3) battery life of 2 h and 9 min (> 2 h).

Regarding the MuscLab dimensions, the final length, width, and height of the case containing the electronic board were 5.4 cm (< 5.5 cm), 4.3 cm (< 4.5 cm), and 2.2 cm (< 2.5 cm), respectively. The flexible textile band presented a length and a width of 33.5 and 18.0 cm, respectively, allowing the monitoring of shank segments with perimeters ranging from 33.5 to 48.7 cm.

The mass of the case and flexible textile band was 40 and 49 g, respectively, comprising a total of 89 g (< 100 g).

TABLE II
DELAY BETWEEN THE MUSCLAB AND EMG DELSYS SIGNALS AND MUSCLAB AND MVN BIOMECH SIGNALS IN MILLISECONDS

Strip	Delsys Trigno™ Avanti			MVN BIOMECH		
	40 bpm	75 bpm	105 bpm	40 bpm	75 bpm	105 bpm
S1	138.0 (82.3)	125.0 (84.1)	134.0 (73.2)	- 5.0 (128.4)	- 4.5 (127.1)	- 8.5 (121.3)
S2	136.7 (71.6)	124.0 (79.3)	134.0 (94.3)	- 8.0 (121.4)	- 3.5 (126.7)	- 4.0 (122.1)
S3	129.0 (66.2)	134.0 (90.7)	120.0 (93.4)	- 66.1 (72.1)	- 69.4 (83.5)	- 60.0 (89.4)
S4	127.0 (92.5)	128.0 (95.0)	113.0 (93.2)	- 76.7 (90.1)	- 61.1 (104.4)	- 67.8 (90.8)
S5	146.7 (68.0)	170.0 (43.2)	178.3 (43.7)	- 38.0 (118.2)	- 46.7 (167.4)	- 32.0 (120.4)
Average (STD*)	135.5 (76.1)	136.2 (78.4)	135.9 (79.5)	- 38.8 (106.0)	- 37.0 (121.8)	- 34.5 (108.8)
Final Average	135.8 (78.0)			-36.8 (112.2)		

*STD means standard deviation

B. Human Tests

Table II presents the results of the MuscLab system's responsiveness. According to Table II, all MuscLab signals were, on average, 1) delayed regarding the EMG Delsys signals (135.8 ± 78.0 ms) and 2) anticipated regarding the MVN BIOMECH signals (-36.8 ± 112.2 ms). Moreover, the average delay of the MuscLab signals measured by each strip was consistent across different cadences, since it varied from 1) 135.5 and 136.2 ms, regarding the EMG Delsys signals and 2) 34.5 and 38.8 ms regarding the MVN BIOMECH signals.

The results of repeatability tests focus on the comparison of the piezoresistive textile strips number 1 and 2 (S1 and S2, respectively, of Fig. 1) with TA EMG signals for dorsiflexion movements. On the other hand, for plantar flexion movements, the piezoresistive textile strips numbers 3–5 (S3, S4, and S5, respectively, of Fig. 1) were compared to GAL EMG signals. In this context, the TA, S1, and S2 signals revealed close average variations of muscle contraction $3.17\% \pm 11.7\%$, $2.97\% \pm 4.32\%$, and $2.41\% \pm 3.77\%$. Moreover, the GAL, S3, S4, and S5 signals revealed an average variation of $4.88\% \pm 13.7\%$, $5.30\% \pm 8.98\%$, $3.47\% \pm 8.68\%$, and $-20.2\% \pm 94.9\%$.

In relation to the magnitude test results, we found an R^2 of 1) 0.91 ± 0.12 and 0.92 ± 0.08 between the dorsiflexion ankle angle and the S1 and S2 strips, respectively, and 2) 0.92 ± 0.15 , 0.85 ± 0.18 , and 0.69 ± 0.22 between the plantar flexion angle and the S3–S5 strips, respectively.

Furthermore, Table III presents the r values obtained between the MuscLab and EMG Delsys signals. The results presented in Table III indicated very strong correlations between the MuscLab signals at S1–S4 strips and EMG Delsys signals, reporting average r values above 0.78 ± 0.08 (p -value < 0.05).

In detail, very strong correlations were achieved between 1) TA signals and both S1 and S2 signals ($r = 0.78 \pm 0.07$) and 2) GAL signals and both S3 and S4 ($r = 0.78 \pm 0.09$ and 0.77 ± 0.07 , respectively).

The S5 signals were moderately correlated with the GAL signals ($r = 0.38 \pm 0.23$). Moreover, low standard deviation

TABLE III

SPEARMAN CORRELATION COEFFICIENT BETWEEN THE MUSCLAB AND EMG SYSTEMS ACROSS PARTICIPANTS. A p -VALUE BELOW 0.05 WAS VERIFIED

ID	Shank Perimeter (cm)	Delsys Trigno™ Avanti				
		S1	S2	S3	S4	S5
1	42.1	0.84	0.83	0.80	0.76	0.33
2	38.2	0.70	0.74	0.77	0.84	0.48
3	38.8	0.86	0.88	0.58	0.71	0.49
4	37.9	0.74	0.71	0.66	0.74	0.75
5	35.0	0.84	0.87	0.87	0.63	0.75
6	36.5	0.67	0.70	0.81	0.77	0.22
7	37.4	0.74	0.75	0.79	0.83	0.17
8	37.9	0.70	0.72	0.81	0.75	0.41
9	35.3	0.88	0.85	0.89	0.84	0.07
10	37.2	0.83	0.71	0.77	0.86	0.15
Average (STD*)	37.6 (1.9)	0.78 (0.07)	0.78 (0.07)	0.78 (0.09)	0.77 (0.07)	0.38 (0.23)

*STD means standard deviation

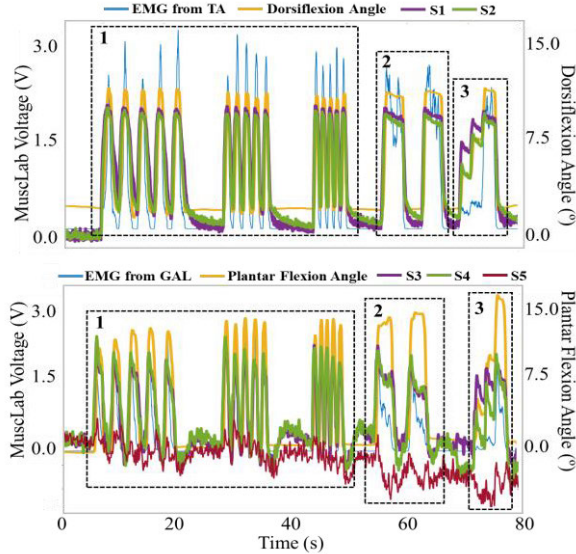


Fig. 6. Participant 2 performing dorsiflexion (top view) and plantar flexion (bottom view) movements. Boxes 1, 2, and 3 represent the responsiveness, repeatability, and magnitude tests, respectively. For easier visualization, the magnitude of EMG signals was scaled to MuscLab signal magnitudes using a factor of $\times 10,000$.

values were verified for S1–S4 (<0.09), while S5 presented the highest standard deviation value (0.23).

Additionally, Fig. 6 depicts the EMG Delsys, MVN BIOMECH, and MuscLab signals for a random participant (participant 2) executing the validation protocol, i.e., responsiveness, repeatability, and magnitude tests.

IV. DISCUSSION

This study proposes a flexible and elastic textile band instrumented with piezoresistive textile sensors to monitor the muscle contraction of individuals with different anthropometries. In an attempt to address the state-of-the-art limitations, the proposed system advances 1) studies [7], [8], [10], [11], [12], [14] by monitoring muscle contraction of different muscle groups simultaneously; 2) study [15] by discriminating the

muscle contraction of agonist and antagonist muscles with a single sensor; and 3) studies [16], [17] by monitoring the muscle contraction of individuals with different anthropometries due to the elastic nature of the proposed solution.

In terms of responsiveness performance, the MuscLab signals were delayed regarding EMG signals (an average delay of 135.8 ± 78.0 ms). According to the study [10], there is a time delay between the onset of EMG and the onset of force generation, defined as the electromechanical delay. This delay typically varies between 9 and 130 ms [25]. Since an FMG, when positioned in a muscle, measures the response of a force generation, its measured signals are expected to be delayed with respect to the EMG signals. In this context, the average delays reported by the MuscLab system (135.8 ± 78.0 ms) are in line with what would be expected because the MuscLab system is an FMG sensor; it, therefore, detects muscle contraction only after the muscle electrical activation.

Conversely, an average delay of -36.8 ± 112.2 ms was found between the MuscLab and MVN BIOMECH signals. Despite the high standard deviation value, these results suggest that the MuscLab signals may be anticipated regarding ankle joint kinematics. These results may be justified by the fact that muscle contractions occur 20–100 ms before the user's lower limb kinematic motion [25]. Furthermore, results indicated that the average delay of the MuscLab signals remained consistent across cadences. These results are aligned with those presented in the study [21] since from a comparison between 16 e-textiles, the Shieldex Technik-tex P130+B (used in the MuscLab system) showed the quickest responses between relaxing and stretching movements.

In addition, MuscLab proved to be repeatable when measuring similar muscle contractions. Considering the repeatability tests, it was found that 1) the variation in the magnitude of the signals measured in strips S1 and S2 ($2.97\% \pm 4.32\%$ and $2.41\% \pm 3.77\%$) was, on average, close to the variation in the magnitude of the TA signals ($3.17\% \pm 11.7\%$) and 2) the variation in the magnitude of the signals measured in strips S3 and S4 ($5.30\% \pm 8.98\%$, $3.47\% \pm 8.68\%$) was, on average, close to the variation in the magnitude of the GAL signals ($4.88\% \pm 13.7\%$). These results are supported by the study [21], which concluded that the used e-textile presented the highest stability, linearity, and low hysteresis. However, the variation obtained for the S5 strip ($-20.2\% \pm 94.9\%$) may indicate that the S5 strip could benefit from a system redesign.

Moreover, the results obtained from the magnitude tests indicated a strong linear relationship between the MuscLab signals at 1) S1, S2, and ankle joint dorsiflexion movements (0.92 ± 0.10) and 2) S3, S4, and ankle joint plantar flexion movements (0.89 ± 0.16). The results are in line with those achieved in studies [26], [27], in which the relation between the muscle bulge of the forearm muscles and the wrist joint angles can be explained by linear regression ($0.92 < R^2 < 0.98$).

The MuscLab system revealed very strong correlations between TA and both S1 and S2 signals and between GAL and both S3 and S4 ($r = 0.78 \pm 0.08$). These results are as expected since the cross-sectional area of a muscle increases

as it contracts, i.e., it shortens and thickens [28]. As the muscle thickens, the MuscLab strips over the muscle are stretched and muscle contraction is detected. Moreover, the standard deviation of the r values for S1–S4 was less than 0.09, indicating that the MuscLab signals were consistently correlated with the EMG Delsys signals across participants. On the other hand, moderate correlations between S5 and GAL signals ($r = 0.38 \pm 0.23$) were found. The higher standard deviation values at S5 suggest that the correlation between S5 and the EMG signals of the GAL muscle varied more than the remaining piezoresistive textile strips.

The results suggest a possible redesign of the MuscLab system. The similar performances of S1 and S2 with TA signals indicate that the system could be simplified to a single piezoresistive strip for monitoring TA muscle contraction. Regarding the S3–S5 strips, different assumptions need to be made. While in some individuals, the S3 strips revealed higher metrics with the GAL activity; in other individuals, the S3 metrics were overtaken by the S4 metrics. However, it was not found a relation between the achieved correlations for S3 and S4 and the shank perimeter. Thus, both strips should remain in the current design of the system. On the other hand, despite obtaining moderate performances in some tests, the S5 strip can be removed as it provides no additional information about GAL activity.

Overall, the MuscLab system successfully detected muscle contractions at different motion cadences, ensured signal repeatability, and distinguished contraction levels across ankle joint angles. Its flexible and elastic properties support usage across diverse anthropometries. While the absence of an analysis of contractions over time and sensor sensitivity is a limitation, future validation tests are planned to address these gaps. In addition, the integration of multiple sensor types, as demonstrated in the study [19], may be a promising way to extend the system to other applications. To this end, future improvements include exploring the fusion of MMG and FMG sensors, by using the IMU capabilities of the MuscLab system for improving muscle contraction monitoring and for joint and segment angle estimations. Additionally, adopting a matrix configuration with flexible piezoresistive textiles, as reported in [16], may enhance muscle contraction discrimination across body segments.

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