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Original Research Article

Characterization of muscle fatigue in the lower limb by sEMG and angular position using the WFD protocol

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A B S T R A C T

Lower limb muscle fatigue has been evaluated in previous studies to understand pain-related movement variability by analyzing different muscles using surface electromyography (sEMG) and angular position signals; however, further studies are needed to particularly understand strength loss due to gait and to inform the development of intelligent control systems for rehabilitation devices in the prevention and management of musculoskeletal or balance control disorders in the Latin American population.

A pilot study was developed to characterize muscle fatigue using a walking fatigue detection (WFD) protocol, an instrumented orthosis and a treadmill. Electrical activity was acquired from Rectus Femoris (RF), Biceps Femoris (BF), Tibialis Anterior (TA) and Gastrocnemius Lateralis (GL) muscles, as well as the angular position of the hip and knee of sixteen healthy Latin-American women, aged 22–34 years, 63.5 ± 6 kg mass, and 161 ± 7 cm height. Data were analyzed with a one-way ANOVA analysis of variance and Tukey's test. Preliminary results show that muscle fatigue is clearly identifiable and is represented by a decrease in both amplitude and frequency of the sEMG signal and lower limb angular position. Muscle fatigue was evident in 93.75% of the participants at the end of the test. 75% of the participants experienced muscle fatigue halfway through the test, of which 31.35%

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were unable to regain strength causing more muscles to fatigue, due to the extra effort they were enduring it was also found that when one muscle goes into fatigue, another muscle supports the action observing muscle compensation but without a uniform pattern.

1. Introduction

Loss of strength due to excessive exercises of the limbs is known as muscle fatigue [1]. It is reflected in decreased muscle electrical activity and the joint motion range, which can affect the gait physical aspects [2] such as coordination, stability and balance. Different sensors have been used to measure these characteristics, but accelerometers and sEMG signals are the most reported, with different positioning strategies and acquisition methods [3].

Currently, muscle fatigue of the lower limbs has been evaluated to know the variability of movement, related to pain, by analysing different muscles [4] of people (healthy or with some disability) with common physical characteristics [5–10], through wearable sensors in the lower limbs to give feedback [11], and allow its use in control strategies of rehabilitation devices or leg orthoses, which allow smoothing and assisting movements during rehabilitation therapies [12 13] in people with some related pathology [14].

Many muscles are needed to generate coordinated lower limb flexion–extension and abduction–adduction movements [15]. Muscles provide strength to the body, generating contractions, activations and movements, so each muscle has a specific function. For the hip, the maximum movement ranges for flexion and extension are usually between 145° and 90° respectively. For the knee, flexion has a maximum movement range close to 135° and for hyperextension a maximum movement angle of −10° [16]. During gait, the angles of movement are small, but many muscles of the lower limbs intervene to generate the necessary muscle contractions to maintain body balance.

Authors have used different protocols to analyse muscle fatigue processes, for example, muscle fatigue in athletes has been extensively evaluated to identify the muscles involved during prolonged exercise and those that fatigue first. This fatigue is reflected in decreased muscle electrical activity and athlete speed [2], for sports simulators. Sit-to-stand exercise without using the hands is one of the most common protocols for inducing fatigue in lower limb muscles. The study presented reviews muscle fatigue under normal gait conditions in common subjects not classified as athletes, this in order to evaluate the pattern of muscle electrical activity during prolonged periods of gait for use in compensation algorithms that use EMG as feedback.

A pilot study to characterize muscle fatigue in the lower limb using a walking fatigue detection (WFD) protocol is shown. The study evaluates the electrical muscle activity of the Rectus Femoris (RF), Biceps Femoris (BF), Tibialis Anterior (TA) and Gastrocnemius Lateralis (GL) muscles and the angular position of the hip and knee. The main objective of this study is to determine the fatigue time, based on maximal voluntary contraction (MVC) [17] and muscle activity of Latin American young women between 22 and 34 years of age without mobility problems in the lower limbs. Each participant walked on a treadmill for half an hour at a constant speed of 4.5 km/h using an instrumented lower limb orthosis. Signals acquired from the different sensors integrated into the orthosis were analysed with an one-way analysis of variance (ANOVA) and Tukey's test to determine the variability of lower limb flexion and muscle electrical activity over time.

The results of this multidisciplinary study have potential utility in the development of intelligent controllers that take into account the electrical activity of the patient's muscles as a method of feedback, since, as demonstrated, muscle fatigue is represented by a decrease in both the amplitude and frequency of the sEMG signal and the angle of the lower limb, which precludes working with fixed trajectories. However, the muscle compensation mechanism was observed in each of the individuals in the study, which, although not conclusive due to the variability presented, indicates a study trend being necessary to record a longer time window and perform a greater number of trials in the study population, as well as to increase the size of the study population.

2. Materials and methods

2.1. *Study participants*

Participants from the Centro de Investigación en Ciencia Aplicada en y Tecnología Avanzada (CICATA), Queretaro, Mexico were selected, by inclusion criteria and following the requirement ethical protocol. A group of sixteen healthy women between 22 and 34 years old, with a mean ± standard deviation (SD) of 63.5 ± 6 kg mass and 161 ± 7 cm height were considered. Participants reported their healthy condition and had no previous injuries and surgeries in at least six months. Inclusion criteria were women over 18 years of age; exclusion criteria were people using technical support for walking, pregnant, previous surgery on lower limbs, pathologies and gait diseases.

Most of the participants do physical exercise every week, as shown in Fig. 1. Each participant was asked about the level of physical activity to find out if the group of people in the study was sportively active, finding that there is an average of 3 days of physical activity per week. This means that they have a similar physical condition among them.

Participants were informed about the purpose of this study, ensuring that personal information would be protected, referring to each participant's information with a number, e.g. S1, S2, . . ., S16. In addition, each participant gave

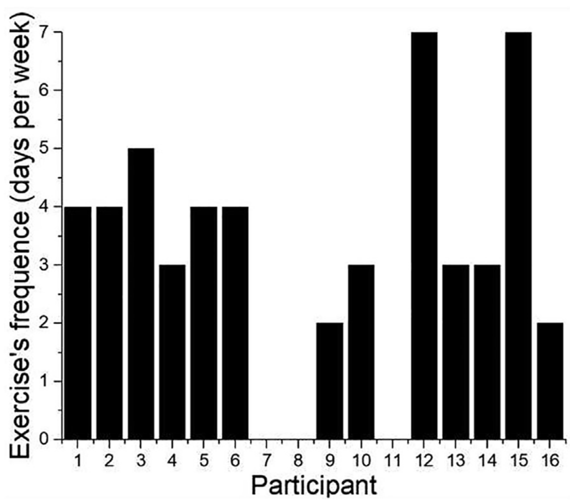

Fig. 1 – Exercise's frequency per week reported by the participants.

prior written consent to participate in the study. The study conformed to the Declaration of Helsinki [18] and local regulations.

2.2. *Acquisition and normalization of sEMG signals*

The orthosis used in the study is instrumented with a MyoWare muscle sensors (AT-04–001) [19] to obtain muscle electrical activity. These sensors are specially designed for robotics, medical devices, wearable electronics and prosthetics/orthotics applications. The muscle activation measurements, performed in differential mode, are represented by a voltage signal. The sensor provides the envelope of the sEMG signal, passing previously through an amplifier, rectifier and finally through the integrator (the EMG envelope). The gain potentiometer was set to 50KΩ, as recommended by the manufacturer. The electrodes used were 32x38 mm disposable self-adhesive monitoring electrodes (AMBIDERM) for pediatric use, for use without additional conductive gel, with an inter-electrode distance of 10 mm.

The data were normalized from 0 to 1 to eliminate uncertainty with respect to the micro-volt scale. For this purpose, the maximum voluntary contraction (MVC) of each muscle by isometric contractions was used. It has been reported in [20] and [21] that depending on the anatomy of the muscle, exercises are derived to obtain the maximum pressure on them and record the MVC of each particular muscle. Generally, unilateral flexion and/or extension exercises are performed using body weight and a support surface to induce resistance to muscle movement. Participants performed the recommended exercise for 5 s, pausing for 60 s, as recommended in [21]. The action was repeated 5 times for each participant.

2.3. *WFD experimental protocol*

The WFD experimental protocol uses a right lower limb orthosis that was designed and constructed from a post-operative knee brace and some parts printed on a 3D printer with PLA material. The lower limb orthosis is instrumented and provides information about hip and knee angles. To place the precision potentiometers in the orthosis, the mechanical elements for mounting the sensors were fabricated using a 3D printer. For the hip angle, the waist was the reference axis, and for the knee angle, the thigh was the reference axis. Prior to acquiring the angular data, the precision potentiometers were characterized and the angular information was obtained.

To acquire the information from the six sensors, an acquisition system is developed and placed on the lower back. The system has a LiPo battery, a voltage regulator at 3.3 V, an STM32 microcontroller and a Bluetooth system as shown in Fig. 2.

Participants could leave the experiment at any time, as participation was voluntary, as they were informed.

Table 1 shows the characteristics of the main elements of the acquisition system.

A cleaning protocol is necessary to place the muscle sensors [17]. The limb must be shaved to ensure contact of the electrode with the skin, cleaned with cotton and alcohol, and the electrode is placed on the motor point of each muscle to be analysed. Finally, the reference was placed on the ankle, as suggested by SENIAM [20]. To ensure that the electrode remains in the same place, due to sweat or lack of adhesive, an adhesive cloth is placed around it.

The selected muscles (Fig. 3) have functions in two joints and when contracted can generate movements in both joints [22].

The RF muscle contribute to hip flexion, flexion of the leg over the thigh, and knee extension [23]. The BF muscle contributes to lateral rotation of the hip and knee, and knee flexion. In addition, the longer portion of the BF extends the thigh over the pelvis [24]. The TA muscle [25] stabilizes the ankle at ground contact during stance phase, generates vertical stability of the leg, and generates dorsiflexion (toe-off elevation) during gait to prevent toe drag. Finally, gait propulsion resides in the GL muscle, becoming the main motor, it also causes the foot plantar flexion of the foot and marginally contributes to leg flexion [26].

Each participant walked for half an hour on a treadmill at a constant speed of 4.5Km/h. Each participant was asked to walk normally, as if the lower limb orthosis was not in place. Thirty minutes of the experiment were selected so as not to generate alterations in the measurements, with the WFD protocol, and to be able to observe the normal muscle pattern in the participants. The participant started walking when the treadmill was set at a speed of 4.5 km/h and approximately two steps later, from the right leg, the transmission of the information from the sensors to the PC was activated. To perform the statistical analysis of the participants, the angular information was taken into account to update the database so that all the participant's signals started at the same phase.

Heart rates was monitored at the beginning, at 15 min and at the end of the test. Likewise, with a smart watch, the number of steps was monitored, as shown in Fig. 8. The mean number of steps of the participants was 2231 during the half hour.

Participants were scheduled from 8:00 to 10:00 a.m. to perform the experiment and maintain a signal acquisition proto-

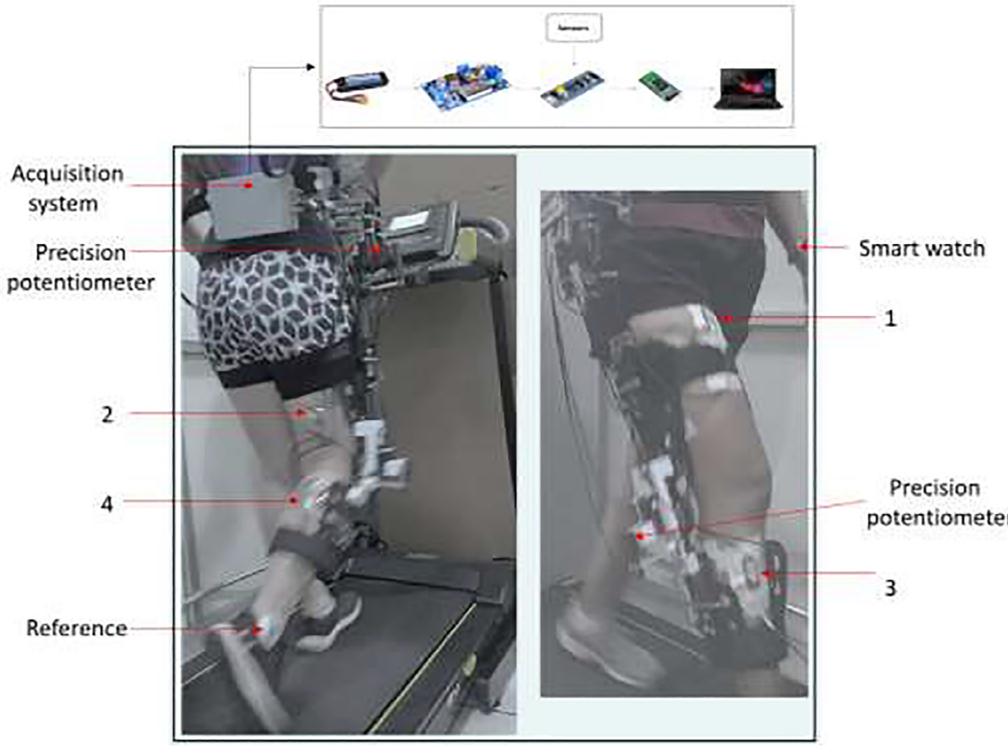

Fig. 2 – Placement of the sensors on the body and on the lower limb orthosis, where 1, 2, 3, and 4 are the MyoWare' sensors placed on the muscles.

Table 1 – Main instruments of lower limb orthosis.

Device	Characteristics
MyoWare	Muscle sensor with filters and amplifier integrated in it
6187R1KL1.0	One-turn linear precision potentiometer
Treadmill	Commercial
Electrode	(AMBIDERM) of 32x38 mm for pediatric use

col that was not affected by physical and/or nutritional variables.

At the end of the test, the electrodes are removed and discarded, each participant's information is stored in the database, and the lower limb orthosis is maintained and cleaned before being used by the next participant.

2.4. Data analysis

To analyse the participant's muscle pattern, the variables time and muscle data at the beginning of the test, after fifteen minutes and at the end (after 30 min of testing) were used. Data from approximately five steps, identified with precision potentiometers placed in the joints, were considered.

The variation in signal amplitude is analysed by repeated measures ANOVA, with a significance level set at 0.05 in order to corroborate the initial hypothesis.

To identify which group or groups of data (e.g., RF 1S, RF 15MIN, and RF 30MIN) make the difference during the test, the TUKEY TEST was used, in which the means of each group of data are pooled and compared to each other to see if one of these samples differs significantly from the others. An alpha level of 0.05 in the statistical analysis was used.

3. Experimental results

Muscle pattern analysis has long been studied to identify behaviours, motor problems, evaluate the evolution of patients against treatments among others [27], showing that although muscle patterns differ for people of different ages, physical characteristics, and each person walks differently [28,29], it is possible to regroup participants with similar final statistical results.

3.1. General analysis of the participant group

In this experiment, the maximum amplitude values of each participant during the test were considered, and three-time instants were considered to analyse acquired data: at the start (1S), at 15 min (15MIN) and at 30 min (30MIN) as shown in Table 2.

Fig. 3 – Electrodes positioning in differential mode with a single reference in ankle (Ref). (1) RF, (2) BF, (3) TA, and (4) GL muscles.

The criterion used is a threshold of sEMG signal amplitude, considering a 20% decrease of the signal. This percentage is determined based on the results obtained in the experiment, where 80% of the participants present significant changes in the initial signal. Perhaps, compared to other studies, it is a relatively high percentage, but in this study, it is considered an adequate value to determine that the signal decrease is due to fatigue and not to a voluntary variation in the participant's gait.

It can be detected that during walking the muscle that generates the more muscle activity is the RF which fatigued at 15MIN for the 56.25% of the participants, and the TA muscle the least. For most participants, a pattern of muscle electrical activity is observed even though the muscles are fatigued. In any case, muscle fatigue is related to low signal amplitude.

Joint movement ranges vary for each participant regardless of the number of steps. Information is collected in six different groups:

1) One person in the middle of the test decreases the angle of movement in both hip and knee joints, and then manages to overcome this fatigue by increasing the angle of movement again.
2) In four participants, muscle activity increased up to middle the test, but in the last minutes, fatigue reduces the range of joint movement.
3) Two participants increased the angle of movement as time progressed.
4) In four participants, muscle fatigue increased as the joint movement angle decreased more and more as the experiment progressed.
5) Three participants decreased the movement angle and maintained it after the middle of the test.
6) Two people increased the movement angle and maintained this angle until the end of the test.

Based on the information acquired from the statistical analysis, One-Way ANOVA, for 43.75% of the participants, all the muscles have a $p < 0.05$, so it is necessary to use alternative hypotheses, where at least in one instant of time, the signal obtained is different from the previous one. 6.25% of the participants have three alternative hypotheses and a null hypothesis, where for a specific muscle there is no significant variation of the signal in the three-time instants, and another 6.25% have no significant variation of the signals. For 25% of the participants there is no significant variation of the signal in the three-time instants for one muscle. Finally, for the

Table 2 – Identification of muscle fatigue at 15 and 30 min of test.

Subject	Fatigued muscle at 15 min	Fatigued muscle at 30 min	No fatigued muscle
1	RF	BF, TA	GL
2	RF, BF TA	TA, GL	–
3	RF	GL, RF	BF, TA
4	TA, GL	GL	RF, BF
5	TA, GL	BF, TA, GL	RF
6	RF	BF, TA, GL	–
7	–	GL	RF, BF, TA
8	GL	RF, BF, GL	TA
9	RF, BF, TA	GL	–
10	–	RF, BF	TA, GL
11	–	GL	RF, BF, TA
12	RF, GL, TA	BF	–
13	–	RF, BF, TA	GL
14	RF	GL	BF, TA
15	RF	TA, GL	BF
16	RF	–	BF, TA, GL

remaining 18.75% of the participants, there are two null hypotheses and two alternative hypotheses.

The standard error of the mean (SEM) for 100% of the participants is < 5%, indicating that, even considering more steps information, this would not change the result.

The purpose of using Tukey's test is to identify if there is a significant difference in the three-time instants at a significance level of 5% for 100% of the participants. Each time instant for each muscle can present a variability in the data obtained (blocking variable). Therefore, it is important to consider the same number of steps for each of them. This reduces the risk of any of the measurements being affected by the amount of data analysed.

For the RF muscle, 43.75% of the participants show a significant difference in the three comparisons; another 18.75% of the participants do not show a significant difference between RF 1S and RF 15MIN signals. 25% of the participants do not have a significant difference between the signals. 6.25% of the participants have a significant difference between the signals obtained at RF 30MIN and RF 15MIN. Finally, 6.25% of the participants had no significant difference between RF 30MIN and RF 15MIN signals.

In the analysis of the three- time instants for the BF muscle, 37.5% of the participants do not present a significant difference during the first 15 min of the test. 12.5% of the participants have no significant difference in their signals during the whole test, while another 37.5% of the participants have a significant difference during the experiment. 6.25% of the participants show a significant difference when comparing data from the first 15 min of the test, while for another 6.25% of the participants this significant difference occurs during the last 15 min of the test.

For the TA muscle, 37.5% of the participants present a significant difference between the three- time instants, while another 25% of the participants do not present a difference between their data during the half hour. 12.5% of the participants do not show a significant difference between TA 15MIN and TA 1S data, and another 12.5% between TA 30MIN and TA 15MIN. 6.25% of the participants show a significant difference during the last 15 min of the experiment. Finally, another 6.25% showed a significant difference during the first 15 min.

For the GL muscle, 37.5% of the data presented significant differences during the half hour, while 31.25% did not present any significant difference in their data during the experiment. 6.25% of the participants had no significant difference during the first 15 min. 12.5% of the participants show a significant difference in the data comparison, except between GL 30MIN and GL 15MIN. Another 12.5% of the participants have a significant difference during the first 15 min. For RF muscle, 68.75% of the participants recorded the following pattern: muscle signal with normal gait pattern, fatigued muscle signal (decreasing signal amplitude), normal muscle signal (increased amplitude). For the BF muscle activity pattern, its amplitude is maintained or increased in the middle of the test and shows fatigue at the end of the test due to the extra effort at MIN15, this pattern is observed in 68.75% of the participants, where a neuromuscular compensation is being generated. The TA muscle presents increasing fatigue during the test for 50% of the participants. The GL muscle presents an identifiable and variable gait pattern for the participants. However, 43.75% of the data obtained from this muscle show the same pattern as the TA muscle.

There is a variation in the maximum amplitude values used by the participants throughout the test to perform the movement and generate the displacement, as can be seen in Fig. 4. In the boxplots, the median location of the data can be seen, and the highest maximum amplitude values used during the test are defined, identifying the minimum and maximum values obtained from the muscle signals RF, BF, TA, and GL. In addition, it is observed that at least one person used a large force percentage, reaching values very close of her MVC, due to muscle fatigue, calling extreme values. For other participants, the values are very close to the whiskers, called outliers. In Fig. 4a and b, for the RF y BF muscles, respectively, as the test time elapses, more force is needed to generate the movement. In Fig. 4c, the TA muscle attempts to perform the same force throughout the test, finding symmetry between the data. In Fig. 4d, for the GL muscle, the force is greater in the middle of the test than at the end.

The electronic equipment, sEMG sensors and other experimental conditions could cause some excitement or nervousness in the participants and thus affect the results during the experiments performed in this work. In addition, it takes a couple of minutes to adapt to walking while wearing special sensors or electronic equipment. In Fig. 5 at the beginning of the test (1S), the heart rates of the 16 participants are slightly higher than those at 15MIN, we consider that this is due to the excitement of the participants when wearing a lower limb orthosis (which allows them to walk freely) and adapting to the weight of the orthosis. In the middle of the test (15MIN) the participants' heart rate decreased, making their gait as normal as possible. In the last minute (30MIN) the participants' heart rates increased considerably, possibly representing physical fatigue. However, during all three-time instants, the heart rates remained in the normal beats per minute (bpm). With this information, it is possible to monitor and ensure that participants are not being exposed to exertion outside the normal range [30].

3.2. Analysis of a participant's muscular pattern

A participant is selected to exemplify the analysis of the information obtained in the three-time instants mentioned above.

The hip and knee movement angles help identify whether lower limb fatigue is present. During the test, both angle amplitude and time vary. Fig. 6a and b show the variation of the angles at the hip and knee joints, respectively. Between the initial signal and the 15 M signal, a decrease in angle is observed, also the steps are a slightly faster. Seconds before the end of the test, the angle of movement of the two joints is observed to increase, however the step is still faster than in the previous two moments. Body fatigue can be reflected in different ways, even when the participant does not perceive it (she doesn't feel tiredness or pain).

Fig. 7a, b, c, and d show the data mean for RF, BF, TA and GL muscles in the three instants of time, respectively, through line plots with error bars, which measure the dispersion of the data. The whiskers in these plots represent the lower and upper limits at which the variability of a participant's sEMG signal amplitude lies, and the probability that this

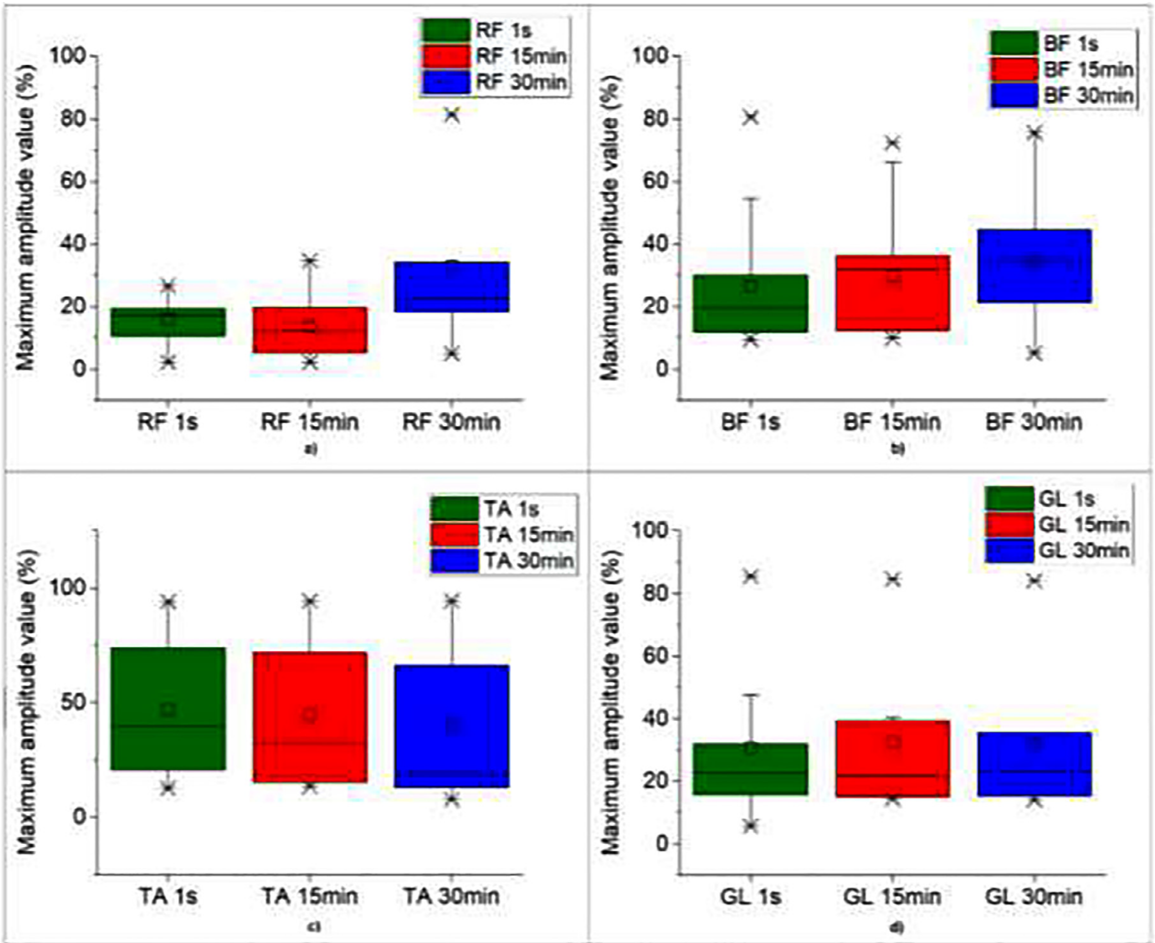

Fig. 4 – Maximum amplitude values used by each muscle: a) RF; b) BF; c) TA; d) GL muscle.
*** represents the single data points obtained during the test, considering extreme values.**

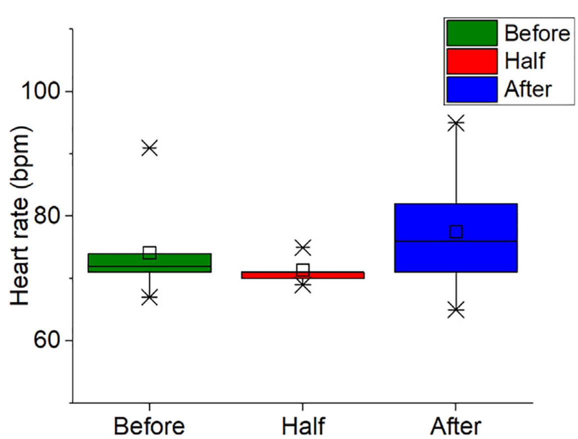

Fig. 5 – Heart rate variation during the test.

amplitude will remain in this values range if the experiment is performed again on another participant. The SE allows us to calculate a confidence interval for the mean, each time we repeat the test, obtained in a sample of the same size from the same population. The variation for the GL muscle is due to the fact that the null hypothesis has not been rejected, this means that all the data means are equal, there is no significant variation in the amplitude of this muscle during the whole test.

For the RF and BF muscle, at the beginning of the test, the SD values are very close to the data mean, implying that the muscle behaves normally. Halfway through the test, the SD values indicate muscle fatigue, moving away from the data mean at that time. At the end of the test, there is a significant variation between the mean of the data and the SD, but it can be predicted that the muscle is recovering from the fatigue phase, since the mean value has increased. In the case of the TA and GL muscles, the mean and SD values are very close in the three-time instants. These values allow us to deduce that during the half hour there was not a sufficiently significant variation in the muscle pattern, indicating that the muscle entered fatigue.

The three-specific signal patterns identified in the participants are shown in Fig. 8. The sEMG signal in black, is the initial signal obtained from the RF muscle of a given participant. The signal during the test, shown in red, represents muscle fatigue, presenting a decrease in both amplitude and time. In this particular case, the electrical activity of the RF muscle is compensated by that of the BF muscle. Finally, the blue-colored signal refers to an sEMG signal in a post-fatigue phase. In this phase, the RF muscle overexerts (recovery

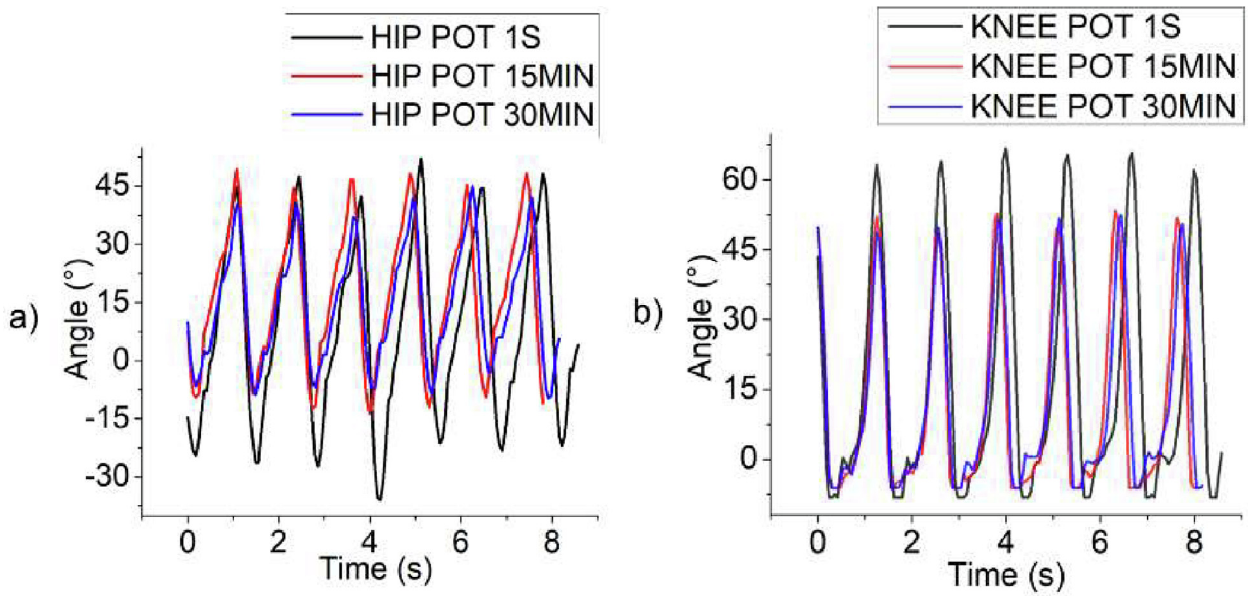

Fig. 6 – Movement angles in the three selected times of a participant: a) Hip and b) knee angles.

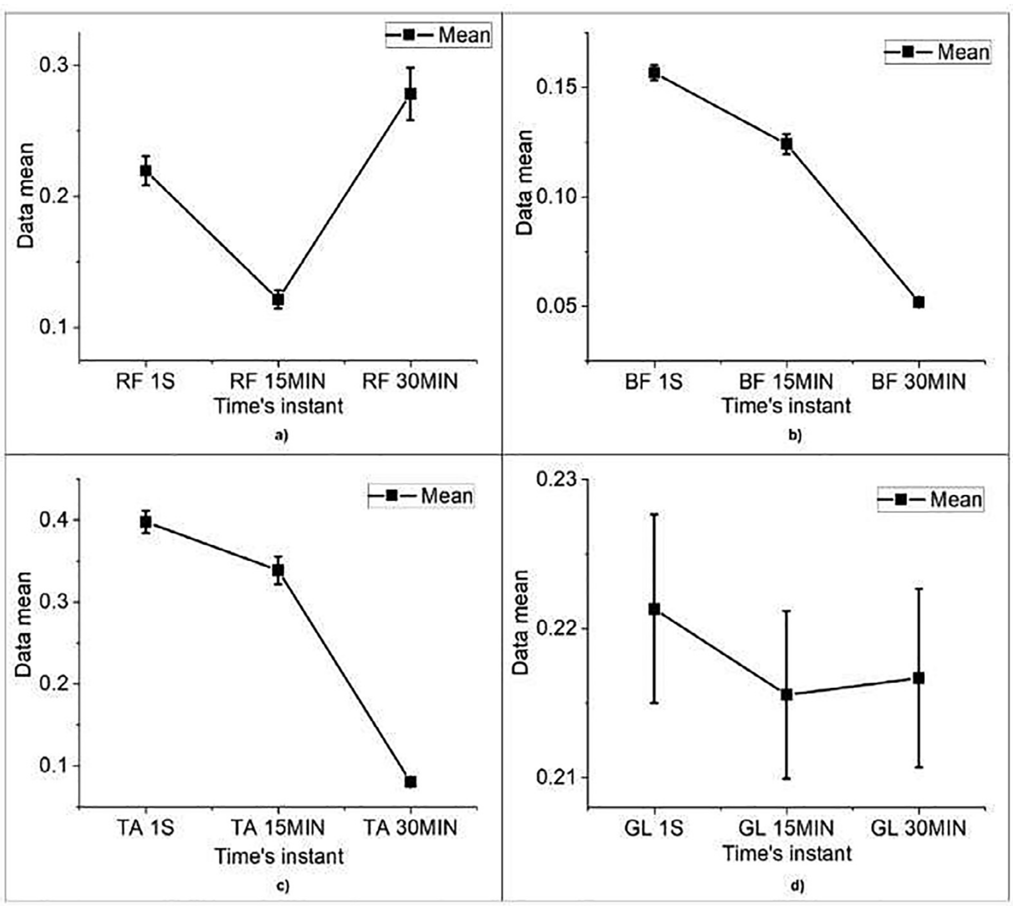

Fig. 7 – Mean (±SE) representing the variability of a participant's data for: a) RF, b) BF, c) TA, and d) GL muscle, using SE as the error.

mechanism) to continue generating movement and helping the muscle that was compensating during the RF muscle fatigue, so the amplitude is higher than the initial one as shown in the green dotted lines.

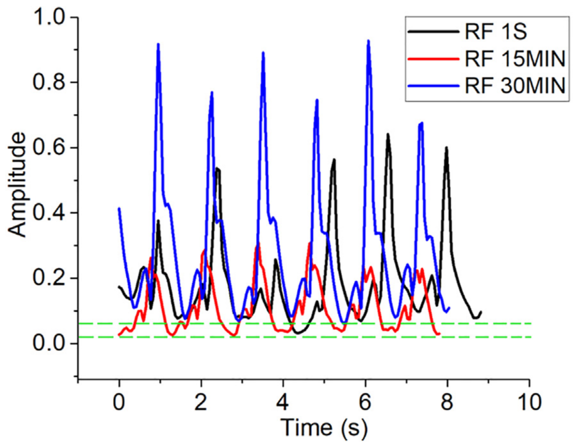

Fig. 8 – Pattern of muscle activity in the three-time instants. In green, with a dotted line, the two baselines are marked, both for the normal and fatigued signal, as well as for the post-fatigued signal.

4. Discussion

In the WFD protocol implemented, the electrical muscle activity and angular position of the right lower limb of 16 healthy Latin American women with similar physical characteristics were used, analyzing the gait characteristics during a controlled 30-minute walk. It was possible to confirm that each participant has a unique pattern during gait, as it is influenced by physiological factors [31].

In this study, 75% of the participants were found to have a minimal muscle fatigue at the middle of the test, and by minute 30, 93.75% of the participants has muscular fatigue.

Halfway through the test, the most fatigued muscle is the RF muscle (biarticular muscle), which was the case in 56.25% of the participants. At 30 min, the muscle that presented the greatest fatigue was the GL muscle, this being the case for 62.5% of the participants. The muscles that fatigued the least, for this experiment, are the BF and TA muscles, with 43.75% of the participants.

Some muscles of 31.25% of the participants went into fatigue halfway through the test and failed to regain their strength at any point, causing more muscles to fatigue at the end of the test, due to the extra effort they were enduring.

The muscle behavior patterns found with this pilot study were similar to the results of other studies. In [32,33] it is indicated that in muscles such as the rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA) and gastrocnemius (CG) there is a decrease in muscle electrical activity, which can be compensated by an increase in the activation of synergistic muscles. In the case of the study, synergistic muscles were not analyzed, so a decrease in the electrical activity of the muscles used was observed.

During the pilot test, a variation in muscle electrical activity and in the angular position of the lower limb was identified as being related to fatigue. Thus, when muscle fatigue occurs in one of the muscles, this force must be compensated in one or more synergistic muscles involved in gait by appearing by neural compensation strategy, to maintain balance and body displacement [2].

A decrease in the electrical activity of the TA and BF muscles was clearly identified during the test. The RF and GL muscle, on the other hand, showed a decrease at 15MIN denoting fatigue, then had a post-fatigue behavior increasing the electrical amplitude until even exceeding the initial level, as occurred with RF. The velocity was constant, so there is evidence of compensation between the muscles involved, i.e., when a muscle decreases its electrical activity amplitude that force is compensated by another muscle as shown in Fig. 8. In this sense, the pattern of muscle fatigue and when it recovers its strength, is similar for all participants.

It is important to consider information from muscle signals when gait analysis is to performed, in order to know how our body behaves internally, regardless of whether there is any previous injury or pathology. This information could prevent future injuries [34].

In rehabilitation, these analyzes make it possible to identify and determine muscle function in each gait's phases or cycles, both qualitatively and quantitatively [35]. In the post-operative patients' case or with certain pathologies, this type of tool provides relevant information to the therapist on the evolution of the patient and the muscular effort made when following a trajectory.

For controlling upper and lower rehabilitation devices, the results found are very relevant. The signal amplitude, in general, can determine how much a robotic device should assist during therapy [36]. When the amplitude is at zero, the muscle activity pattern is normal, therefore, the motors are off. As the amplitude of the signal decreases, this is when muscle fatigue begins to be observed, and this is when the motors should assist in recovery therapy. When the amplitude is greater than the initial amplitude, and is greater than zero, the motors must continue to assist the movement of the participants, as there is muscle overstrain.

For athletes [34], understand how their muscles behave while training would allow them to determine the time and amount of exercise they should perform before their muscles become fatigued or injured.

The placement of the electrodes [37,38] and the sweating of the participant during the test are considered relevant factors in data acquisition, as they may alter or modify the actual information coming from the muscles. However, the electrode placement protocol was performed as reported. The authors consider it relevant to select a specific schedule or range for the experiment, trying to standardize parameters that may affect signal variation in different participants, e.g., hydration level, food intake.

5. Conclusion and future research

In the pilot study presented, the results obtained in 16 healthy female participants from Latin America are analyzed from the muscle pattern and angular position of the right lower limb before, during and after muscle fatigue. A decrease in both amplitude (signal intensity) and step size are detected, as well as a pattern of recovery from muscle fatigue. It is noteworthy that muscle fatigue is rapidly evident and was identifiable in all participants, on the other hand for the muscle compensation mechanism of fatigue was observed in different muscles.

The mechanism of muscular compensation of some muscles with others seems to avoid a body imbalance, since a constant speed was maintained. Despite having a limited sample size, the results can provide information on muscle fatigue. However, the authors consider that in order to conclude about muscle compensation mechanisms it is necessary to record a longer time window and to perform a larger number of trials in the study population, as well as to increase the size of the study population.

In this type of study, a series of fundamental parameters of the participating population should be considered, among which are that the population should be healthy with similar physiological characteristics, weight, height and age should be considered, as well as not having a history of muscular dystrophies or biomechanical alterations, in addition to taking into account the physical activity they have, to determine repetitive or similar gait patterns as the time of the test elapses. However, instrumented orthoses are also useful for acquiring not only angular gait information, but also muscle behavior using sEMG.

This study has potential use as a basis for using the sEMG as feedback in adaptive controls that allow the active participation of patients by monitoring their pattern in real time, taking into account that the rehabilitation devices that currently exist, are based on tracking trajectories at the same speed and with the same force, and the effect of fatigue should be considered, since it affects the initial trajectory, on the other hand it is necessary to continue studying the mechanisms of muscle compensation, to consider this effect also in mechanism-assisted rehabilitation therapies. It should also be considered that the results obtained can be used in athletes or people who are not yet injured, but have aspects to work on in their gait.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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