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STUDY OF ELECTROMYOGRAPHIC PATTERNS OF ERECTOR SPINAE AND LOWER-LIMB MUSCLES DURING DIFFERENT MODALITIES OF GAIT IN POST-STROKE INDIVIDUALS

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Ph.D. Thesis presented to the Postgraduate Program in Biotechnology (RENORBIO) at the Federal University of Espirito Santo as a partial requirement to obtain the degree of Doctor in Biotechnology.

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"The nitrogen in our DNA, the calcium in our teeth, the iron in our blood, the carbon in our apple pies were made in the interiors of collapsing stars. We are made of starstuff."

Carl Sagan

ABSTRACT

Stroke is one of the leading causes of motor disability in the world. New technologies have been developed to increase efficiency and reduce costs of rehabilitation of post-stroke individuals. **Objective:** To compare electromyographic patterns related to muscle onset/offset, duration of activation and analysis of neuromuscular fatigue of erector spinae (ES) and lower-limb muscles during different modalities of gait in post-stroke and healthy individuals. **Methodology:** The changes in the median frequency (MDF) was analyzed during isometric tasks and walking on a treadmill in healthy individuals (N = 10) to identify fatigue. Ten post-stroke and 30 healthy subjects participated of the second stage of the study, in which ES and three lower-limb muscles were analyzed during different gaits (walking on treadmill and ground, with and without arm swing, and using a walker), with the neuromuscular fatigue analyzed in stroke gait. Muscle analysis was also conducted with two post-stroke subjects while using the UFES's robotic walker. **Results:** For the healthy subjects, all the lower-limb muscles showed reduction in their MDF during walking on treadmill. Walking on treadmill had a stronger influence on the onset/offset muscles than the arm swing in the healthy individuals. For post-stroke subjects, their ES muscles presented a similar pattern to the healthy subjects, but the contralateral side had longer activation near the toe-off than the ipsilateral side in both gaits. All the observed changes in the activation for each phase indicated a longer duration of activation of the post-stroke subjects. Regarding neuromuscular fatigue, it was not possible to detect reduced MDF values for post-stroke individuals. The use of the UFES's robotic walker improved the symmetry of one post-stroke subject, and the symmetry of duration of activation in the swing phase for all muscles of the other subject. **Conclusion:** MDF changes were detected in non-strenuous exercises in healthy subjects. ES muscle activation is not influenced by arm swing in healthy individuals, with the same behavior in post-stroke individuals. As a finding of this research, we concluded that trunk muscles can be used in rehabilitation processes and also to control robotic devices for assistance or rehabilitation.

Keywords: Erector Spinae; Gait; Neuromuscular Fatigue; Stroke; Trunk Muscles.

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SUMMARY

1. INTRODUCTION

Stroke has been considered the main cause of neuromuscular damages worldwide (BELDA-LOIS et al., 2011; WHO, 2015) and it is the second most common cause of death in the world, with around 12.5% of all the deaths (WHO, 2015). Stroke is characterized as a neurological deficit attributed to an acute focal injury of the brain by a vascular cause, including cerebral ischemia, intracerebral hemorrhage or subarachnoid hemorrhage (SACCO et al., 2013).

In this brain attack, the subjects' independence is reduced because they are not able to perform many daily tasks, such as walk, feed themselves or dress up, which results in physical, psychological and economic problems. Hemiparesis (partial loss of movements on contralateral side to the lesion), muscle spasticity and poor balance (resulting in difficult to walk, and risk of falls) are some of clinical features in post-stroke individuals (CAPÓ-LUGO; MULLENS; BROWN, 2012). Most of post-stroke subjects need rehabilitation, mainly aiming to independence improvement, for instance gait recovery, and independence in basic tasks (ROGER et al., 2011). An incomplete recovery not only maintains the abnormal pattern of the paretic limb, but can also impair the contralateral limb, due to the constant presence of compensatory mechanisms during gait, causing secondary complications because of mobility decreased (ALLEN; KAUTZ; NEPTUNE, 2011; MILOVANOVIĆ; POPOVIĆ, 2012).

Development of technologies for post-stroke gait rehabilitation, to increase training efficiency and reduce costs has been an important subject in the literature (EDELSTEIN, 2013; HELAL; MOKHTARI; ABDULRAZAK, 2008; SHEFFLER; CHAE, 2015). Robotic devices can provide higher efficiency, precise movements and greater repeatability in rehabilitation tasks. For example, robotic exoskeletons can allow enough flexibility (mechanic and control) in the joints of the subject's lower-limbs, in order to train movements in daily activities like walking, upping and down stairs, sitting down, etc. (SHEFFLER; CHAE, 2015).

For the command of these technologies, surface electromyography (sEMG) has been the most common technique used to infer motion intention of users. Usually, researchers have used sEMG signals captured from the user's lower-limbs, mainly from flexor and extensor knee muscles, as they are directly related to the desired motion (DAWLEY; FITE; FULK, 2013; FLEISCHER; REINICKE; HOMMEL, 2005; HARGROVE et al., 2011; HUANG; KUIKEN; LIPSCHUTZ, 2009; KIGUCHI; IMADA, 2009).

Trunk muscles have recently arisen as an alternative to control robotic devices (DELISLE-RODRIGUEZ et al., 2015), as their activity may be more preserved in stroke (DICKSTEIN et al., 2004). On the other hand, the capture of signals from trunk muscles is more comfortable, there is the additional possibility of assessing the subject's posture during the rehabilitation sessions, and the signal activity may anticipate propulsive phases in gait with a cycle pattern (DELISLE-RODRIGUEZ et al., 2015). However, few studies have used trunk muscles for use in gait analysis (CECCATO et al., 2009; WHITE; MCNAIR, 2002) and in robotic exoskeletons (DELISLE-RODRIGUEZ et al., 2015).

On the other hand, when robotic devices are used in rehabilitation, it is possible to have longer, more precise and greater repeatability (EDELSTEIN, 2013). However, the chance of the subject having neuromuscular fatigue increases. In this case, the exoskeleton would not have utility, as the subject would not be able to continue the task until he/she has recovered (XU; CHU; ROGERS, 2014).

Thus, this research deals with the investigation of the following electromyography features of trunk and lower-limb muscles: muscle onset/offset, duration of time activation and changes in median frequency, which can provide information necessary to aid in the assessment of the rehabilitation process of the subject, and also in the development and evaluation of robotic devices for rehabilitation.

1.1. OBJECTIVE

1.1.1. General Objective

Analyze and compare the electromyographic patterns of erector spinae and lower-limb muscles during different modalities of gait in post-stroke and healthy individuals, which can contribute for post-stroke rehabilitation.

1.1.2. Specific Objectives

- Compare the activation of erector spinae (ES) and lower-limb muscles in five different modalities of healthy gait;
- Analyze the influence of the modified conventional walker and treadmill gait in the ES and lower-limb muscle activation during healthy gait;
- Verify the influence of arm swing on ES activation during healthy and stroke gait;
- Identify the neuromuscular fatigue during isometric exercises and gait at normal speed, using the short-time Fast Fourier Transform (STFFT), in healthy individuals;
- Compare the activity and the fatigue of trunk muscle and lower-limb muscles during free and walker-assisted gait in post-stroke individuals;
- Verify the activation symmetry of muscle function of ES and lower-limb muscle in post-stroke individuals during free and walker-assisted gait;
- Identify the neuromuscular fatigue over time during free and walker-assisted gait in post-stroke individuals;
- Analyze the influence of the UFES's robotic walker on the muscle activation, speed and duration of gait phases in post-stroke individuals.

1.2. THESIS STRUCTURE

In the Chapter 2, a literature review is presented, which approaches stroke and its sequelae, mainly in the hemiparetic gait, but also addressed robotic devices for stroke rehabilitation and gait analysis.

The Chapters 3, 4 and 5 consist of articles developed during this Ph.D. Thesis, which were submitted to journals in the field of this research. The first article is a study of activation of trunk and lower-limb muscles in five gait modalities of 30 healthy individuals, and compares the influence of treadmill, walker and arm swing in the muscle pattern. This article was written with the goal of explain how trunk muscles act in the healthy gait, and presenting data to compare with post-stroke individuals. In the second article (Chapter 4), sEMG signals obtained from 10 subjects during isometric tasks and gait on treadmill were analyzed through short-time Fast Fourier Transform (STFFT) with the objective of detecting neuromuscular fatigue in non-strenuous exercises. The importance of this article was to test a protocol and the STFFT technique to detect fatigue in post-stroke individuals. Chapter 5 presents and article with the main focus of this research, which was a study with 10 post-stroke individuals, who performed free and assisted gait using a modified conventional walker in which was possible to assess muscle onset/offset, duration of activation in stance and swing phase, symmetry between both contralateral and ipsilateral sides, and neuromuscular fatigue. In the Chapter 6, case studies with two post-stroke individuals using the UFES's robotic walker are presented, in which a similar experimental protocol described in Chapter 5 was used.

Chapter 7 consists of the description of our patent application, which was developed in partnership with colleagues of the NTA-UFES. This patent of product is composed of two modules, in which the first sends position information to the second module, which is used to synchronize inertial and biological signals, as sEMG. This patent has arisen from the need to use different equipment of acquisition of biologic signals, which must be synchronized to biomechanics data, with extremely importance for gait analysis.

Finally, Chapter 8 presents the final considerations about this research, showing also the contributions and publications achieved, in addition to future works.

2. LITERATURE REVIEW

2.1. STROKE

Stroke is one of the most common causes of walking disabilities, with approximately 60% of the individuals suffering from persistent problems in walking (VAN KAMMEN et al., 2017). After the acute phase, between 20% to 30% of affected individuals are unable to walk, and most post-stroke individuals have gait difficulties, such as reduced speed or dependence on the use of assistive devices (BUURKE et al., 2008; MA; CHAN; CARRUTHERS, 2014).

Stroke is characterized by sudden blood deprivation to a specific region of the brain, which may be of ischemic or hemorrhagic origin (Figure 1), leading to neuronal death at the affected site (WHO, 2011). Ischemic stroke is the most common type, accounting for 85% to 90% of cases. Although less common than ischemic stroke, hemorrhagic stroke has a higher mortality rate, ranging from 33% to 45% (HOLLANDER et al., 2003; OVBIAGELE; NGUYEN-HUYNH, 2011), and often results in greater functional impairment (ALAWIEH; ZHAO; FENG, 2018).

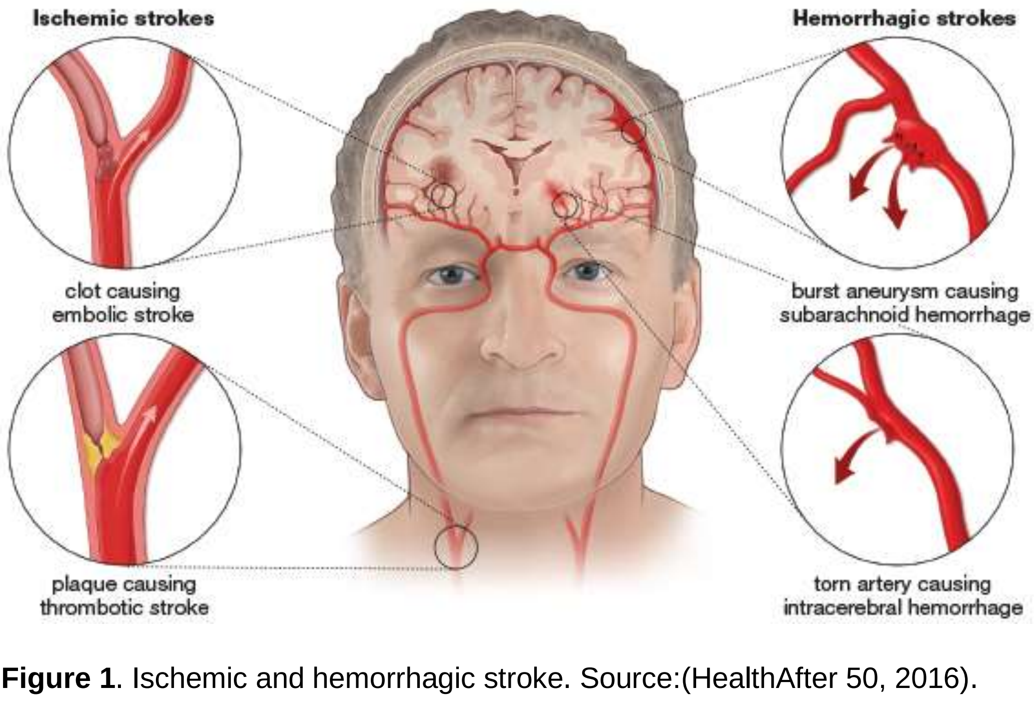

Figure 1. Ischemic and hemorrhagic stroke. Source:(HealthAfter 50, 2016).

2.2. SEQUELAE

The severity of a stroke, regardless of the type, varies according to the extent and location of the brain in which the injury occurred (DEB; SHARMA; HASSAN, 2010). The clinical signs and impairment resulting from stroke are directly related to the brain area that has been damaged (BELDA-LOIS et al., 2011). For example, a lesion in the corticospinal tract above the pyramidal decussation, in case of the carotid, middle cerebral or anterior cerebral arteries have been injured, can cause decreased motor ability and muscle weakness, occurring mainly in the body part contralateral to the lesion (BELDA-LOIS et al., 2011; MACHADO; HAERTEL, 2014). On the other hand, lesion to the anterior cerebral artery (Figure 2) causes contralateral hemiparesis, with predominance of the lower limb, thus affecting the ability to walk (PARE; KAHN, 2012).

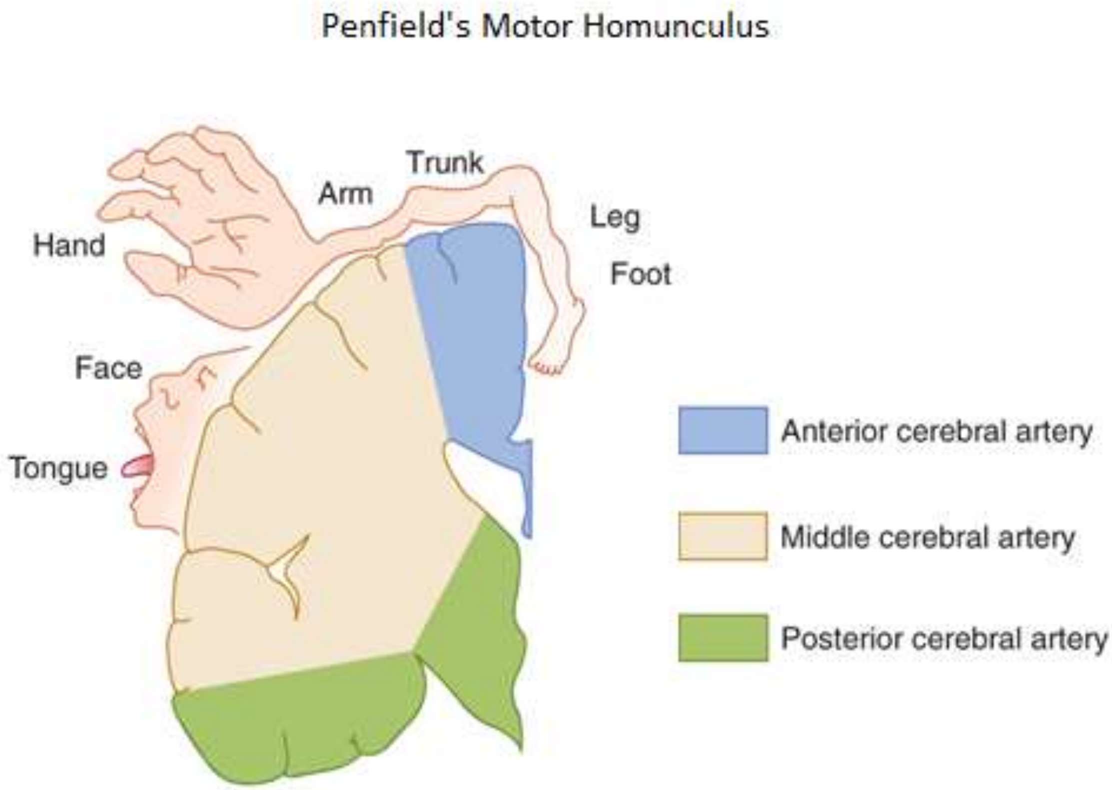

Figure 2. Penfield's Motor Homunculus, which shows a representation of the areas and proportions of the human brain dedicated to processing motor functions, the brain area irrigated by the anterior cerebral artery is shown. Source: (WAXMAN, 2013).

Initially, motor disabilities after stroke occur solely as a direct consequence of the disruption of descending neural pathways from the brain cortex to the muscles, caused by neuronal death, with no direct lesion on the musculoskeletal system (CORBETTA et al., 2015; LUI; NGUYEN, 2018). Nevertheless, structural changes in skeletal muscles can arise over time after stroke (BERENPAS et al., 2017) such as, for instance, atrophy in the contralateral side muscles that is result, mainly, of the muscle disuse (SCHERBAKOV; SANDEK; DOEHNER, 2015). Also, fatty infiltration, fibrous tissue (RYAN et al., 2011; SCHERBAKOV; SANDEK; DOEHNER, 2015) and changes in aerobic capacity (BUURKE et al., 2008) are commons in these muscles after stroke. However, Berenpas et al. (2017) found that these changes are not restricted to the muscles on the contralateral side, being that both sides showed deviations in comparison to reference values from healthy subjects. Regarding the loss of muscle strength, studies conducted by (DORSCH; ADA; CANNING, 2016) have identified that contralateral and ipsilateral lower-limbs present, in average, between 48% and 66% of the muscle strength of healthy participants.

As a consequence of stroke, the individual can present abnormal timing and amplitude of muscle activation, and impairment in the mobility and stability of joints, in addition to spasticity (exaggerated tonic reflex resulting in sudden spasmodic muscular movements), muscle weakness and impaired postural control (CAPÓ-LUGO; MULLENS; BROWN, 2012; VAN KAMMEN et al., 2017). These damages interfere in simple daily tasks, as they hinder stability, mobility, balance, and walking (BRUNI et al., 2018).

2.3. HEMIPARETIC GAIT

Quadriceps femoris (vastus medialis, vastus lateralis, vastus intermedius and rectus femoris) and triceps surae (gastrocnemius medialis, gastrocnemius lateralis and soleus) muscles are spastic in those patients whereas hamstrings (biceps femoris, semitendinosus and semimembranosus) and tibialis anterior are flaccid, hindering the knee flexion and dorsiflexion (MURRAY et al., 2014; SHEFFLER; CHAE, 2015).

Researches carried out by (DORSCH; ADA; CANNING, 2016) analyzed maximum isometric strength of contralateral lower-limb muscle groups of 60 post-stroke individuals, finding they were weaker than that of healthy individuals. For the spastic muscle groups, knee extensors presented 45% of the strength of the control, and ankle plantar flexors, 57%, whereas for the flaccid muscles, knee flexors showed 40% of the strength of the control, and ankle dorsiflexors, 35%.

In spite of flexor weakness, post-stroke individuals present more co-contractions between agonist and antagonist muscles than healthy subjects in order to avoid knee hyperflexion and plantar hyperflexion (SHAO et al., 2009). For instance, due to weakness in knee flexors, such as biceps femoris, there is lower propulsion performed by the contralateral limb (ROUTSON et al., 2013). Nevertheless, these muscles show higher activation time, with coactivation between quadriceps and hamstrings muscle groups (CORRÊA et al., 2005). Similarly, studies of Shao et al. (SHAO et al., 2009) showed that gastrocnemius medialis (plantar flexor) got active during the initial contact, at a time in which this group are usually not active. Both lower-limbs can develop muscle coactivation, being that in the ipsilateral lower-limb it can aid to establish the walking ability (ROSA et al., 2014), however, the coactivation can increase the energy cost associated with locomotion after stroke (LAMONTAGNE; RICHARDS; MALOUIN, 2000).

Thus, post-stroke individuals tend to produce a compensatory movement in order to walk, resulting in the typical hemiparetic gait (Figure 3) that is often characterized by stiff-legged gait (reduced range of knee motion) and drop foot (lack of ankle dorsiflexion during swing), leading to raised hip during swing (YAVUZER, 2006), which is known as hip circumduction (WHITTLE, 2007), shown in Figure 3.

Therefore, compensatory mechanisms result in an asymmetric gait, in which the ipsilateral limb is predominantly used, overloading it and at the risk to produce musculoskeletal injury (ANDROWIS et al., 2018; BEYAERT; VASA; FRYKBERG, 2015). Temporal parameters in stroke gait are characterized by a shorter stance phase and longer swing phase of the contralateral limb than the ipsilateral one, and as higher this asymmetry (contralateral/ipsilateral ratios) as slower the self-selected gait velocity (AWAD et al., 2015; LEWEK et al., 2014). In addition, arm swing and trunk kinematics during stroke gait present asymmetry, which is also result of the spasticity in the upper-limb (JOHANSSON et al., 2014).

Figure 3. Hemiparetic gait. The pictures show a subject with hemiparesis (flexor spastic pattern of the upper limb and extensor spastic pattern of the lower limb), and also shows how the lower limb performs the circumduction to carry out the gait progression. The right figure shows the circumduction, where the contralateral lower-limb in the swing phase moves in an arc, rather than straight forward. Source: modified of (WHITTLE, 2007).

Stroke gait speed varies among different levels of motor damages and time after stroke. Thus, hemiparetic gait has a wide range from 0.10 to 1.00 m/s, according to (BALABAN; TOK, 2014; LAMONTAGNE; RICHARDS; MALOUIN, 2000), and a range from 0.23 to 0.73 m/s, according to (BARROSO et al., 2017; VERMA et al., 2012). Fritz and Lusardi (2009) claim gait speed of more than 0.80 m/s is necessary for effective ambulation in the outside. Although gait velocity is used to predict gait-related motor dysfunctions, in the case of post-stroke patients, compensatory mechanisms may cause an increase in speed, although there is no recovery of normal movement patterns (BARROSO et al., 2017).

Because of this gait asymmetry and lack of balance, about 75% of post-stroke patients need assistance for walking independently during the first three months (VERMA et al., 2012). However, there are no evidence-based criteria for choosing the device to help the patient (VERMA et al., 2012). Tyson and Rogerson (TYSON; ROGERSON, 2009) evaluated the use of cane and foot-ankle orthosis, which provided more confidence and safety to the patients (20 post-stroke patients; time since stroke: 6.5 ± 5.7 weeks), improving their functional mobility.

In the chronic stroke, after 6 months it occurred, is common that the individual can walk, although with assistive devices, however gait and balance problems persist, interfering in the patients' quality of life (BRUNI et al., 2018).

2.4. TRUNK MUSCLES

Whole body is involved in walking, although the lower-limb muscles are the main actuators, whereas trunk muscles provide flexibility and integrity of spine (ANDERS et al., 2007). Trunk muscles have an important function for maintaining the balance, posture and combine anticipatory and reactive actions during walking (CECCATO et al., 2009; KARTHIKBABU et al., 2012; PEREIRA et al., 2011).

Lower trunk muscles maintain a more stable level of activity during sustained limb extension, whereas the upper muscles are more involved in countering reaction forces generated by limb movement onset (DAVEY et al., 2002), controlling the weight shifts and the movement of the trunk against the gravity (KARTHIKBABU et al., 2018).

In addition to provide stability, trunk muscles can also execute movements, which is the case of erector spinae (ES) that is a larger powerful muscle (ANDERS et al., 2007). Actually, the ES (Figure 4) is a group of muscles (spinalis, longissimus, and iliocostalis) located on each side along the spinal column (DE SÈZE et al., 2008). It is considered the main muscle of the back (CIONI et al., 2010).

The ES presents a descending activation pattern during gait (CECCATO et al., 2009; DE SÈZE et al., 2008; KARTHIKBABU et al., 2012). Its activation in normal gait is found around the toe-off, in both sides, i.e., during the two double supports of the gait cycle (CIONI et al., 2010).

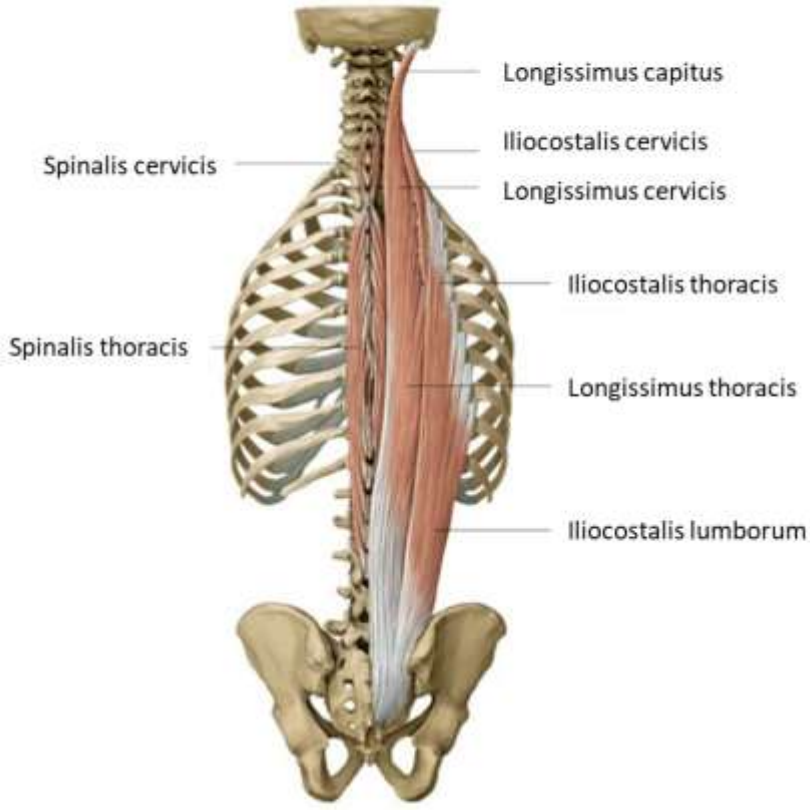

Figure 4. Erector spinae muscle group: cervical, thoracic and lumbar regions. Source: (GILROY; MACPHERSON; ROSS, 2012).

After a stroke, the trunk is bilaterally impaired (VAN CRIEKINGE et al., 2017). Studies conducted by Fujiwara et al. (2001) found that, after stimulation, non-affected hemisphere evoked a bilateral response in trunk muscles. This study showed that the intact hemisphere becomes responsible for restoring the trunk function, likely by potentiating the effects of preexisting uncrossed motor pathways. Therefore, trunk muscles receive bilateral innervation from the motor cortex, and, when compared to the limbs, these muscle impairments are less remarkable (QUINTINO et al., 2018). In fact, trunk impairments are common in post-stroke individuals, reducing balance and trunk coordination (VAN CRIEKINGE et al., 2017).

Buurke et al. (2008) evaluated 13 post-stroke individuals periodically for up to 24 months through sEMG of both ES and lower-limb muscles, using evaluation scales. They found that, there were no significant changes in the muscle activation timing, even with improvement of gait parameters. Thereby, it was supposed that gait improvement is due to compensatory mechanisms of legs and trunk. Other study (MARCUCCI et al., 2007) analyzed ES (L3 level) of 8 post-stroke individuals and 8 control subjects. During maximum isometric voluntary contraction, the ES muscle of the post-stroke individuals had significant differences, with lower activity than the control group in both sides.

For an application in the control of robotic devices, the employment of signals from the ES may be more comfortable, as in using electrodes on the trunk there is the possibility to cover them with clothes, which gives a more natural look, resulting in less psychological and social problems. On the other hand, these muscles are less affected in post-stroke individuals because of its innervation, there is the possibility of having an earlier activation in comparison to the lower-limb muscles, in addition to allow assessing the subject posture during the tasks (KOBETIC et al., 2009; PARETTE; SCHERER, 2004).

2.5. FATIGUE

Fatigue often manifests as both physical and mental lack of energy, which is characterized by decreased functional status (GLADER; STEGMAYR; ASPLUND, 2002). The subject, when has muscle fatigue, feels lack of energy, tiredness and difficulty to make strength. Also, there is a decrease in the ability of performing physical activities (LEWIS et al., 2011). In a post-stroke population of 613 chronic patients, fatigue occurred in approximately 30% of patients (FEIGIN et al., 2012), being that it often interferes with the stroke rehabilitation process (GLADER; STEGMAYR; ASPLUND, 2002).

2.5.1. Neuromuscular fatigue

Neuromuscular fatigue is defined as the diminution of muscle strength generation, which occurs due to the imposition of muscle contractions for a long period of time, great force intensity or high rate of repetitive motion (BOYAS; GUÉVEL, 2011; CHANG et al., 2017).

Several physiological factors can generate muscle fatigue, being of peripheral and/or central origin. In peripheral fatigue, there are changes in the concentration of substances that influence the process of muscle contraction, for example, glucose depletion and adenosine triphosphate (ATP), in addition to accumulation of metabolites that are harmful to organism (GUYTON; HALL, 2011). While in central fatigue, there may be a decrease in the firing rate of motor units compared to the rate at the beginning of muscle activation (BOYAS; GUÉVEL, 2011).

A high degree of neuromuscular fatigue limits some daily activities and can interfere in individual's ambulation ability, even in small distances (HESSE, 2006). In post-stroke individuals, the contralateral side develops a higher level of neuromuscular fatigue than the ipsilateral side, in comparison with healthy individuals after a fatiguing task (BOUDARHAM et al., 2014). Therefore, it can create interferences in their life quality and rehabilitation process (LEWIS et al., 2011).

When the individual has fatigue, the training session must be interrupted in order for the patient to recover and then can continue. However, depending on the duration of each session, the recovery may not be carried out, and the individual is not able to complete it. Furthermore, fatigue may cause pain, afraid to continue the therapy, and performance of wrong movements (XU; CHU; ROGERS, 2014).

It is important to highlight that neuromuscular fatigue occurs gradually according to the motion progress. In that situation, the muscle has its maximum strength decreased, due to the available power reduction to perform the task. However, it is possible the subject has muscle fatigue and even so keeps the motor task (ENOKA; DUCHATEAU, 2008). Nonetheless, if fatigue is not considered and the movement is kept, it may result an accumulation of problems, as the muscle does not have enough time to recover itself.

2.6. REHABILITATION

The restoration of motor functions after stroke is a complex and multifactorial process that depends on the severity of the injury, the intrinsic spontaneous recovery of each individual, and the effects of therapeutic interventions (DOHRING; DALY, 2008; ROGER et al., 2011). Most of post-stroke individuals need rehabilitation, whose main goal is to recover movements, allowing them to carry out daily tasks independently (DOHRING; DALY, 2008; ROGER et al., 2011), and recover the walking ability (AGUIAR et al., 2018).

The brain has the capacity to generate new connections to relearn functional movements lost after a stroke; therefore, the rehabilitation process is based on this neural adaptation (ANDROWIS et al., 2018; PALMER et al., 2016; XU; CHU; ROGERS, 2014). In the first 3 months, there is a higher tendency of spontaneous recovery, but the patient may recover functional movement skills also in the chronic phase (BRUNI et al., 2018).

Some authors (ANDROWIS et al., 2018; BEYAERT; VASA; FRYKBERG, 2015; VAN KAMMEN et al., 2017; WALLARD et al., 2015) have approached the task specific repetitive training, based on motor learning and neuroplasticity, considered the main rehabilitation method to recover the functional gait. Overground gait training with assistance, such as parallel bars, is the most common method of clinical practice for post-stroke patients (JETTE et al., 2005). Increases in the gait speed are often during the rehabilitation process, and the gait symmetry uncommonly is improved, therefore, the differentiation between movement pattern recovering and compensatory movements has been considered in studies about stroke rehabilitation (BEYAERT; VASA; FRYKBERG, 2015).

In many cases, the recovery is inefficient, causing a worsening in the clinical status and damage in the ipsilateral limb, leading to decreased mobility and secondary complications (ALLEN; KAUTZ; NEPTUNE, 2011). In addition, conventional gait trainings and rehabilitation methods currently used do not provide a complete restoration of motor function for most patients (DOHRING; DALY, 2008; MEIJER et al., 2011). For this reason, robotic devices have been extensively studied, aiming to be a new rehabilitation strategy for people with severe motor impairment, such as is the case of post-stroke individuals (BELDA-LOIS et al., 2011).

2.7. ROBOTIC DEVICES

Many studies (ANDROWIS et al., 2018; DOHRING; DALY, 2008; NAM et al., 2019; ONEN et al., 2014; WALLARD et al., 2015) have used robotics devices for motor rehabilitation, in order to recover important features of the gait and maintain muscle integrity. It has been shown that the use of robots in rehabilitation leads to a better result, being able to recover characteristics important of the gait. In addition, rehabilitation through robotic devices brings the benefits of being more intensive (high repetition), controllable and motivating; there is also the possibility of quantifying the individual's performance, reducing effort for therapist and healthcare costs (BELDA-LOIS et al., 2011; BRUNI et al., 2018; MEHRHOLZ; POHL, 2012).

In rehabilitation, it is important that the individuals stay active throughout the process to restore their remaining muscle strength, as this will bring the benefit of a faster recovery, with a lower risk of healthy limb involvement and muscle disuse related complications (PENNYCOTT et al., 2012). Therefore, in addition to providing intensive rehabilitation and targeted tasks (BELDA-LOIS et al., 2011), it is necessary for robotic devices to safely rehabilitate, use symmetrical gait patterns, be restorative, and use a physiological pattern similar to physiological activation (SCHULER; MÜLLER; VAN HEDEL, 2013).

2.7.1. Body Weight Support

Brain lesions can result in loss of ability of body weight support (BWS), thus, during gait training a reduction of the body weight over the lower-limbs may become the process more efficient (MUN et al., 2014). For stroke rehabilitation, the BWS can decrease the overload in the ipsilateral limb.

Walkers are beneficial to aid individuals with balance problems and gait disturbances (HELAL; MOKHTARI; ABDULRAZAK, 2008). They can also support a great amount of vertical strength, up to 50% of the user weight (WHITTLE, 2007).

Suica and colleagues (2016) analyzed the immediate effect using a rollator walker in 19 healthy subjects (22 to 70 years). They identified a reduced muscle activity of the lower limbs caused by the weight bearing imposed on the walker. Other study (DRAGIN et al., 2014) of 4-week clinical trial (22 subacute post-stroke patients) using a body postural support connected to a powered rollator walker concluded that this device changes the gait speed and balance control significantly when compared to a control group. Finally, Patel, Vaghela and Ganjiwale (2017) assessed the walking ability of 30 acute post-stroke individuals, after 3 weeks of a custom-made physiotherapy program using knee gaiter in a group and suspended walker in other group. They identified improvement in gait symmetry in both groups, using the three-minute walk test (3MWT) and 10 meter walk test (10MWT).

Many robotics devices with BWS in over-ground gait rehabilitation are found in the literature, for instance, omni-directional mobile platforms (MUN et al., 2014; PATTON et al., 2008; TAN et al., 2013), and with lower-limbs assistance, such as the NaTUre-Gaits (LUU et al., 2014).

2.7.2. Robotic devices applied in rehabilitation

Exoskeleton is defined by Herr (2009) as “a device that amplifies or augments the user’s strength and endurance”. In exoskeletons, actuators are used in parallel to the joint and perform the flexion-extension movement. Generally, the selection of the actuators is based on torque values of each joint during walking in healthy normal speed (WINTER, 2009). In the study conducted by Onen et al. (2014), an exoskeleton was used together with a pair of crutches, to aid in the user stability and to reduce the imposed load on joints, during the gait with 14 healthy volunteers. They have found that, with the use of crutches, the weight carried by the lower-limb was decreased by 47.71% on average. Thus, a decreased load over the exoskeleton caused by a support makes possible to reduce the actuator sizes.

Some exoskeletons for gait rehabilitation drive the patient to follow a gait pattern predetermined, in which the individual intention is not considered (ZHANG et al., 2010). When the user’s motor intention is considered, the neuroplasticity can be explored (XU; CHU; ROGERS, 2014), and, furthermore, interaction forces among the user, robotic device and environment could be combined with the motor intention, and applied in the device controller to generate a more natural and efficacy gait, which make the use of those devices more intuitive (TUCKER et al., 2015).

Lokomat® is a famous robotic device for rehabilitation, which consists of a motorized treadmill, a BWS system and two lightweight robotic actuators attached to the subjects' lower-limbs (COENEN et al., 2012). Many studies have used it in post-stroke rehabilitation to analyze improvements in the patients, such as done by Coenen et al. (2012), which analyzed 10 post-stroke patients during gait on the ground and using Lokomat, verifying a lower muscle activity in the lower-limb muscles, suggesting a lower effort in walking, in addition to a higher symmetry between contralateral and ipsilateral muscle activity. Other research (VAN KAMMEN et al., 2017) also evaluated the gait using Lokomat in 10 post-stroke patients. The results indicated a reduction in the abnormal vastus lateralis and biceps femoris muscle activity and an increased temporal symmetry. Furthermore, study conducted by Wallard et al. (2015) verified the kinematic gait parameters of 10 post-stroke individuals before and after a rehabilitation program (four sessions of 30 minutes, per week, during five weeks) using Lokomat. Results indicated there was a sensorimotor retraining, with the device allowing extended periods of exercise and continuous repetition of gait cycles, which improved the locomotor patterns.

Others robotic devices has been also assessed. For example, Androwis et al. (2018) assessed the use of the robotic exoskeleton EksoGT (which includes two powered joints (hip and knee) and a passively sprung ankle joint with adjustable stiffness) in 5 post-stroke patients. The robotic exoskeleton promoted activations of the VL and RF on the contralateral side, more similar to healthy gait. Moreover, studies of Nam et al. (2019) with 40 post-stroke individuals, using the exoskeleton Exowalk (which provides a stable and firm standing ability with little chance of falling) and found an efficacy similar to the physical therapist-assisted gait training interventions.

Regarding robotic walkers, the ASBGo (MARTINS et al., 2014) uses a conventional four-wheel walker, which was modified to include a support base for the upper-limbs.

This device detects possible falls or the user's loss of balance, and employs force sensors in the forearm supports, enabling it to perceive the user's movements while providing better stability. The omnidirectional walker – ODW (TAN et al., 2013) features four wheels and four force sensors on the forearm support. Four healthy individuals simulating motor disability tested ODW in order to evaluate its adaptive controller through the analysis of the loads imposed on the walker and the changes in its center of mass.

In other study, Morone et al. (2016) evaluated the effects of over-ground robotic walking training (therapy of 4 weeks) using a robotic walker (i-Walker) in 44 post-stroke subjects. This robotic walker improved balance, gait stability and reduced falls of stroke subjects.

Lastly, the UFES's robotic walker (ELIAS-NETO, 2013) was used to evaluate knee kinematics patterns of patients with physiotherapy sessions. This robot walker has arms and hands supports and employs a laser sensor to detection of lower-limb distance. The clinical trials showed that the robotic walker reduced the load on the limbs and helped the balance of subjects with moderate osteoarthritis.

2.8. GAIT ANALYSIS

Human gait is a relatively complex and periodic action with repetitive motions that requires the synchronization of the central, peripheral nervous system and muscles to perform fast and complex movements (CHEN et al., 2013; MISHRA et al., 2012).

The analysis of the human gait pattern by phases allows identifying more directly the functional meaning of the different movements generated in the individual joints and segments, making it possible to determine the kinematic and kinetic parameters and muscular activation by comparing them in different phases (TAO et al., 2012).

The gait cycle begins, conventionally, when there is foot contact with the ground, and ends when there is the next ground contact with the same foot. It consists of two sequential and distinct phases called “stance and swing phases” (Figure 5). The stance phase, when the foot is in contact with the ground, can be divided into five sub-phases: initial contact, load response, mid stance, terminal stance and pre-swing. On the other hand, the swing phase, when the foot is advancing, is divided into three sub-phases: initial swing, mid swing and terminal swing (PERRY; BURNFIELD, 2010).

The gait cycle can be divided relating both limbs, being “stance phase” composed of “first double support” (0-10%), “simple support” (10-50%), “second double support” (50-60%), and “swing phase” (60-100%), with their respective duration in the gait cycle (KIRTLEY, 2006; WHITTLE, 2007). The initial contact of the other limb is in the 50% of the gait cycle and its stance phase ends in the 10% of the cycle.

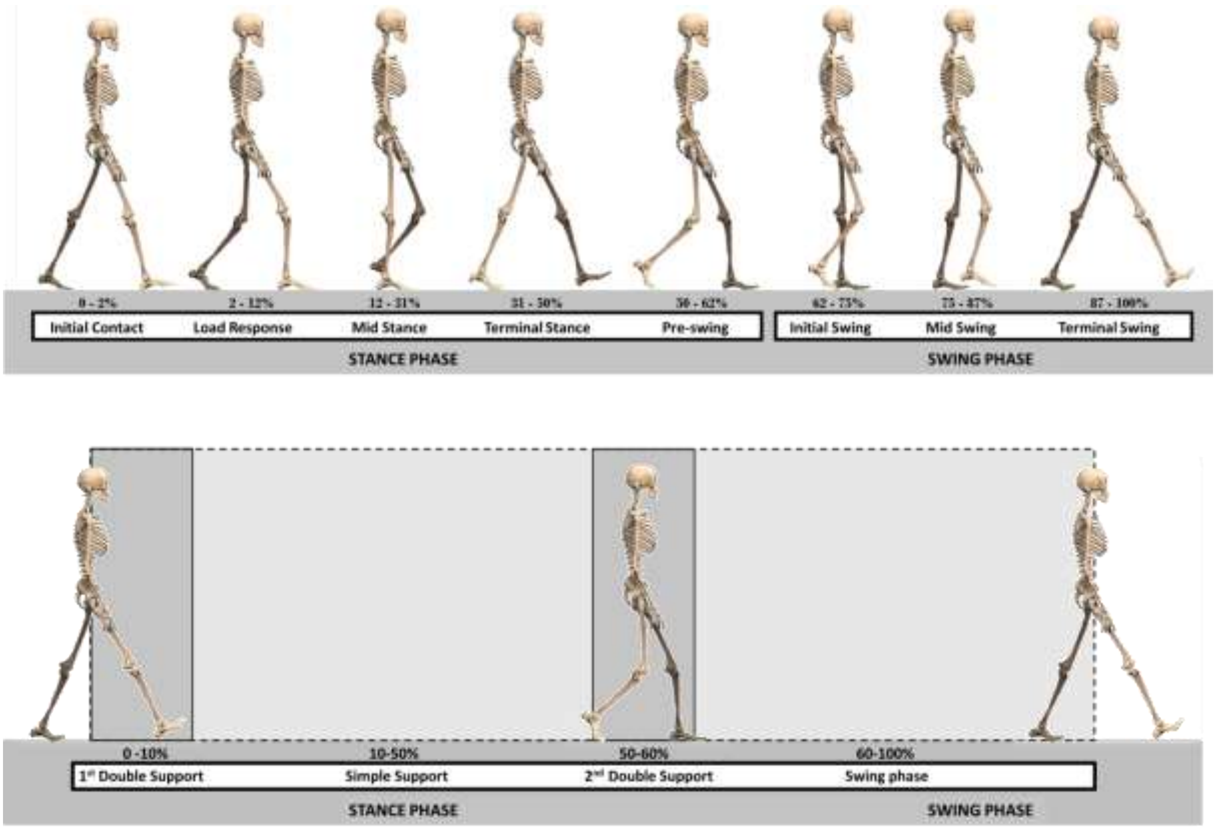

Figure 5. Gait phases. The upper figure shows two main phases (stance and swing), eight sub-phases, and the values of each phase for the healthy gait, according to (PERRY; BURNFIELD, 2010). The bottom figure is divided in stance phase, composed of 1st double support, simple support, 2nd double support, and swing phase, with their respective duration in the gait cycle, such as described by (KIRTLEY, 2006; WHITTLE, 2007).

Gait analysis can be used to characterize walking parameters, determine useful devices for rehabilitation, and to monitor the healing progress of the individual in rehabilitation (TAO et al., 2012). It is a process that involves a group of the following features that can be studied: kinetic, kinematic and electromyographic parameters.

2.8.1. Kinetic

Force platforms are used in order to measure the ground reaction force related to the total effect of load between the subject and the ground. They are considered the golden standard for analysis of gait kinetic parameters (DYER; BAMBERG, 2011), which are, generally, placed on the ground, and have its surface maintained flat, furthermore, they have force sensors that record the applied force on the three axis (BARELA; DUARTE, 2011).

2.8.2. Kinematic

Kinematics describes motion based to position, velocity, and acceleration. It can be measured through photogrammetry, cinematography, footswitches, goniometers and inertial sensors units (accelerometers, gyroscope and magnetometer). Generally, the kinematic parameters of the gait are analyzed by video capture through multi-camera systems, which identify body segments and joint movements. The limitations related to the use of this system of analysis involve the need to be installed in closed environments, preventing its use in outpatient monitoring, and also, implementation costs are quite high when compared to other analysis tools (HAN et al., 2009).

In order to overpass the high cost of the multi-camera systems, some studies evaluated the use of a setup of accelerometers: placed on waist, wrist and both ankles (KHANDELWAL; WICKSTRÖM, 2017); two accelerometers attached to the shoes, one at the level of the heel and one at the level of the forefoot of each foot (BOUTAAYAMOU et al., 2015); upper chest, each anterior thigh, and under each medial forefoot (SAREMI et al., 2006). In these studies, the accelerometry system provided reliable and valid kinematic measurements of the gait.

Actually, the acquisition of kinematic parameters can be done by using only an accelerometer, reducing thus the discomfort of the subject, time and cost of the process. In fact, the use of a single accelerometer on the ankle of subjects allowed estimating their kinematic gait parameters (HAN et al., 2009; LEE et al., 2010). On the other hand, the curve shape from accelerometer has two characteristic peaks, where is possible to divide the gait in two phases: stance and swing phase (LEE et al., 2010). In addition, other studies used only a single accelerometer in the lower back in healthy subjects and elderly with knee osteoarthritis (CLERMONT; BARDEN, 2016; GODFREY et al., 2015).

Using only an accelerometer, it is possible also to analyze both lower-limbs simultaneously, identifying stance and swing phase, from which it is possible to obtain information on double and single support, as well as gait symmetry. In the hemiparetic gait of post-stroke individuals, an important parameter that indicates gait improvement is the reduction of asymmetry. For these individuals, the gait analysis using only one accelerometer can be done on the lower back, making the analysis more practical and allowing the analysis of both legs at the same time.

2.8.3. Electromyography

Electromyography (EMG) records motor unity action potentials during a voluntary muscle contraction, in other words, it determines the electric activity of the analyzed muscle (ROJAS-MARTÍNEZ et al., 2013). The surface electromyography (sEMG) is a simple and non-invasive method, which consists in electrodes fixation on the superficial muscles. In gait analysis, it is an important tool to provide information about the relative contribution of the muscles during the movements (CAMPANINI et al., 2007).

Before starting an electromyographic analysis, it is necessary to know both anatomically and functionally about the musculature involved with the specific movement to be evaluated. In addition, the electrodes choice interferes with the obtained signals. Furthermore, following the recommendations of Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscle (SENIAM, 2016), it is used Ag/AgCl electrodes, using bipolar configuration, with 10 mm diameter discoid format and conductive gel. The Ag/AgCl electrode is indicated for having a stable behavior, presenting low noise. The bipolar configuration also reduces noise, because it has a high common mode rejection rate.

For skin preparation, prior to the electrodes placement, cleaning and subsequent trichotomy of the defined region should be done, which results in an abrasion of the skin with 70% alcohol, to remove dead cells and other skin impurities, which can interfere in the contact between electrode and skin (CRISWELL, 2011; MERLETTI; PARKER, 2004; HERMENS et al., 2000). During the electrode fixation, it is necessary to identify the correct location in which the electrode will be placed, through the initial posture, designated by SENIAM, being this location specific for each muscle. The arrangement of the electrodes may affect the characteristics of the sEMG records (CAMPANINI et al., 2007). Therefore, it is recommended that the electrodes are arranged following the direction of the muscle fiber, and maintaining an inter-electrode distance, defined as the distance between the centers of the conductive areas of each electrode, of 20 mm. Nevertheless, a reference electrode should be placed in a specific region of the analyzed limb where there is no contact with muscle fibers, usually on the ankle, patella or spinous process of the C7 vertebra (SENIAM, 2016).

As for the sampling frequency, the Nyquist-Shannon theorem says that a sampling frequency that is at least twice the maximum frequency of the signal should be used. The myoelectric signal has frequency information up to 500 Hz, so the sampling frequency used in sEMG records must be at least 1 kHz (MERLETTI, 1999).

During the test, there may be interference in the myoelectric signal, resulting from, for example:

- Cable movement: the electrodes and cables should be kept attached to the skin during the all data collection through adhesive tape or elastic. This procedure must be done to avoid possible movement artifacts, caused by cable instability (HERMENS et al., 2000; MERLETTI; PARKER, 2004);
- Crosstalk: One of the main concerns is the occurrence of crosstalk, which is present exclusively in the sEMG. Crosstalk is the interference in the myoelectric signal caused by the activation of muscles adjacent to the analyzed. This interference becomes significant when there is a need to determine the activation time of different muscles, as in the case of motion analysis (MERLETTI and PARKER, 2004). However, crosstalk can be

reduced by the correct size of the conductive area of the electrode, decrease of the inter-electrode distance which limits the surface area under the electrodes and fixation of the electrode on the center of the muscular surface (HERMENS et al., 2000);

- Electromagnetic devices: some electrical and electronic equipment can also generate interference. The main frequency component of the electrical grid, in this case, is 60 Hz. To eliminate this noise, a band reject filter can be used in the range of 60 Hz (WINTER, 2009).

The comparison of the myoelectric signal is hampered by the wide anthropometric differences existing among individuals, and even in the individual, due to the specific characteristics of each body region. Hence, the importance of normalization of the myoelectric signal, which will bring the values of all the signals into percentage values (0-100%), making them possible to be compared (CRISWELL, 2011). There are several ways to normalize the sEMG signal amplitude, such as: voluntary maximum contraction, voluntary submaximal contraction, maximum signal peak during the task, and average signal during the task.

In people with normal neural control, the most convenient reference is the normalization process by sEMG recorded during the maximal effort test (PERRY; BURNFIELD, 2010). However, in gait analysis, voluntary maximal contraction normalization is less reliable than the value obtained from contractions during the performed task. Marchetti and Duarte, (2006) argue that signal peak-to-peak is the best way to normalize dynamic contractions. The peak of the myoelectric signal is particularly applicable for patients with neurological lesions who have suffered damage in voluntary control such as spastic disabilities, which can happen in stroke individuals. They cannot reliably produce a maximum effort for the normalization reference (PERRY; BURNFIELD, 2010). In the case of hemiparetic individuals, the use of maximum voluntary contraction is not indicated, since they have higher rates of use of their maximum voluntary force during walking than healthy people, making comparisons between gaits difficult to be established (LAMONTAGNE; RICHARDS; MALOUIN, 2000).

3. EFFECT OF DIFFERENT MODALITIES OF GAIT ON ERECTOR SPINAE AND LOWER-LIMB MUSCLES ACTIVATION PATTERN

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3.1. ABSTRACT

Purpose: This work analyzes the trunk erector spinae (ES) and lower limbs muscles activation during walking, in different gaits, which is quite useful to get information about rehabilitation progression and posture. Additionally, it compares gender-related features in the gait performance. **Methods:** To develop this study, 30 healthy volunteers aged 18-38 years (15 males and 15 females) were selected. Five modalities of walking were performed by volunteers, being one assisted, and two with a treadmill (with and without arms swing), and two directly on the ground (with and without arms swing). To analyze the gait cycle and muscle activation the volunteers had electrodes placed on specific points in their body, to measure the muscle electrical activity, along with accelerometer to measure the gait stance and swing phases. **Results:** The data indicates erector spinae (ES) muscle has a rostrocaudal sequential activation pattern in the gait cycle, being two periods of activation predominantly in the double support phases. Gait without arms swing does not affect the normal muscle pattern during walking. Walking on treadmill had a greater influence on the onset/offset muscles than arm swing. The duration of the phase stance in the group, and the gait parameters in men and women did not show significant statistic difference. **Conclusions:** As conclusion it was found that the ES muscle activation is not influenced by arm swing, however, it is affected by gait on treadmill. The parameters shown in this work can be used to compare healthy and pathological gait and provide information about the rehabilitation progression of people affected by mobility, and also posture impairments.

Keywords: Assisted Gait, EMG, Erector, Spinae, Lower Limbs, Treadmill Gait.

3.2. INTRODUCTION

Trunk muscles have an important function for maintaining the balance, posture and combine anticipatory and reactive actions during gait (CECCATO et al., 2009; KARTHIKBABU et al., 2012; PEREIRA et al., 2011). The trunk is arranged for different layered muscle groups in order to perform movements, keep the trunk erect and stabilize the body, combining flexibility and stiffness during motions (SWINNEN et al., 2012). In addition, these muscles are responsible for weight transfer between limbs and thorax, and pelvis rotation inversion (WHITE; MCNAIR, 2002). Lower trunk muscles maintain a more stable level of activity during sustained limb extension whereas the upper muscles are more involved in countering reaction forces generated by limb movement onset (DAVEY et al., 2002).

Some studies (CECCATO et al., 2009; DE SÈZE et al., 2008) claim that back trunk muscles are sequentially activated, as the lumbar and thoracic components of the erector spinae (ES) act synchronously. Regarding the different levels of the cervical column, ES activation occurs, firstly in C7 level and further, in the other levels sequentially, ending in L3 level (PERRY; BURNFIELD, 2010).

ES is one important muscle involved in both stability and motion during gait (CECCATO et al., 2009; DE SÈZE et al., 2008), as it is a lengthy muscle, making feasible to analyze it in many levels of the vertebral column. It is covered in lumbodorsal fascia and nuchal fascia, between T5 and T11, and there is no fascial window above it. Because of that, some levels cannot be assessed, due to the crosstalk and noises that disturb the signal quality. Even so, it is the most superficial muscle involved in stability of the body during gait (DE SÈZE et al., 2008).

White and Mcnair (2002) analyzed ES and abdominal muscles during treadmill gait (speed of 4 km/h) in 38 healthy subjects. For ES muscle, three groups of different patterns were found and the main difference among them was the signal amplitude. In all the groups, there were activity peaks near the initial contact phase, and decreased activation was observed in the initial contact of the contralateral limb to the studied one.

Anders et al. (2007) performed an investigation to identify muscle activation patterns of trunk muscles of 15 healthy men during treadmill with a function of walking speed (2, 3, 4, 5 and 6 km/h). Among the analyzed muscles, multifidus (MF) and ES (L1 level) muscles were both characterized by clear phasic activation patterns. MF was little activated during slow walking speeds, but showed increased amplitude peaks at both initial contacts in faster gaits. On the other hand, ES only showed one relevant peak during contralateral initial contact and its activation matched characteristics related to global stabilizing muscles. Corroborating this information, the results found by Zoffoli et al. (2017) for 18 healthy subjects walking on treadmill showed that ES was primarily engaged during the initial contact of the contralateral foot, especially at low speeds.

Arm movements have been included in gait analysis (DAVEY et al., 2002; MEYNS; BRUIJN; DUYSENS, 2013; MULLINGTON et al., 2009). Even so, there is no agreement that arms movement during gait is passive (as consequence of thorax motion and inertia) or driven by muscle activity (MEYNS; BRUIJN; DUYSENS, 2013; MIRELMAN et al., 2015). Arm swing is a characteristic of human walking and running; it is like a pendulum motion in which each arm swings with the motion of the opposing leg, balancing out of phase with our legs (GOUDRIAAN et al., 2014; HUSSEIN; ABD-ELWAHAB; EL-SHENNAWY, 2014; JOHANSSON et al., 2014; PONTZER et al., 2009). In addition, it may minimize energy consumption, optimize both stability and neural performance (HUSSEIN; ABD-ELWAHAB; EL-SHENNAWY, 2014; MEYNS; BRUIJN; DUYSENS, 2013; MIRELMAN et al., 2015). However, during arm swing, stabilizing trunk in response to arm abduction is required, hence some muscles are activated. In fact, some studies found increased activation in trunk muscles in the contralateral side to the abducted arm (DAVEY et al., 2002; MULLINGTON et al., 2009).

In the study performed by Mullington et al. (2009), 19 healthy right-handed volunteers were analyzed, observing ES muscle at T12 and L4 levels and the rectus abdominis at L4 vertebral level. The responses shown by the trunk muscles seemed are dependent on the direction of the arm movement, speed and whether the motion was expected or not.

Because the direct and important relationship with walking, some studies (ANDERS et al., 2007; WHITE; MCNAIR, 2002; ZOFFOLI et al., 2017) have approached the role of trunk muscles in treadmill gait analysis. About gait on the ground, Cecatto et al. (2009) investigated postural equilibrium during gait initiation and walking and Cromweel et al. (2001) studied the mechanism for trunk stabilization. However, as far as our knowledge no studies were conducted about the effect of treadmill walking on trunk muscles compared to walking on the ground. Neither the influence of arm swing on muscle activation during gait was properly addressed, mainly in walker-assisted gait. The aim of this study is to analyze activation of the erector spinae and lower limbs muscles in five different modalities of gait, comparing ground and treadmill gait, assisted and free gait, males and females walking, and the influence of arm swing on ES activation. This study may contribute, for instance, to evaluate the rehabilitation progress of people under therapy and the use of walking assistance devices.

3.3. MATERIAL AND METHODS

3.3.1. Volunteers

Thirty healthy subjects (15 males and 15 females; 27 ± 5 years; 169 ± 10 cm height; 67 ± 15 kg weight; Body Mass Index: 23 ± 4 kg/m^2) volunteered for the experiments. This research was previously approved by the Ethical Committee of Federal University of Espirito Santo (UFES/Brazil), number CAAE: 64797816.7.0000.5542, and all the volunteers signed the informed consent.

Eligibility criteria for inclusion in this study were: be 18 to 59 years old (adult subject); have no motor impairment or pain (in order to no affect the walking); be able to walk on a treadmill; have enough cognitive skills and language for following the experiment instructions. Individuals were excluded if they have had any musculoskeletal or neurological disorder limiting ambulation, and/or if they had cardiorespiratory impairment.

3.3.2. sEMG and accelerometer data acquisition

sEMG and accelerometer data were recorded simultaneously using an acquisition equipment EMG 830C (EMG System do Brasil Ltda®) with 16-bit analog/digital conversion resolution, amplifier gain up to 2000V/V, common mode rejection > 100 dB, input impedance of $10^9\Omega$, and sampling frequency of 1000 Hz.

For comparison, sEMG signals from both right side of the trunk and right lower limb were considered, such as done by White and McNair (2002). The trunk muscle ES (C7, T12 and L4 levels), lower limb muscles (rectus femoris - RF, biceps femoris - BF, and vastus lateralis - VL) were analyzed. After marking two points, 2 and 4 cm laterally from the spinous process in C7 and L4, respectively, the points were joined by a line, and the electrodes were placed on them at each spine level to be studied, such as suggested by De Sèze et al. (2008).

The region around the bipolar electrodes (Ag-AgCl, pre-gelled, 25 mm of inter-electrode distance and 10 mm of diameter) was cleaned and shaved to reduce skin-electrode impedance, and their cables were fixed on the body to minimize motions artifacts. The electrodes positions were determined following recommendations from SENIAM (2016), and according to (DE SÈZE et al., 2008; SWINNEN et al., 2012). A reference electrode was placed on medial malleolus, and the accelerometer (with the y-axis pointing cranially and x-axis pointing anteriorly) was placed above the lateral malleolus.

3.3.3. Experimental Protocol

The volunteers were asked to perform the gait on the ground (Figure 6), with comfortable speed, as follows:

- **(a)** Walk for 10 meters with normal movement of the arms (gait with arm swing on ground - GAS);
- **(b)** Walk for 10 meters without arms movement, i.e., with arms resting on the side of the trunk (gait with no arm swing on ground - GNAS);

(c) Walk for 10 meters using a modified conventional walker with wheels, adjusting the height to maintain the volunteer's body erect (assisted gait on ground - AG).

The items **(a)**, **(b)** and **(c)** were performed three times each, with intervals when necessary. Furthermore, the subjects walked on a treadmill at a fixed speed of 1.0 m/s for 3 minutes:

(d) With arms swing (gait with arm swing on treadmill - TAS);

(e) Holding on the treadmill support (gait with no arm swing on treadmill - TNAS)

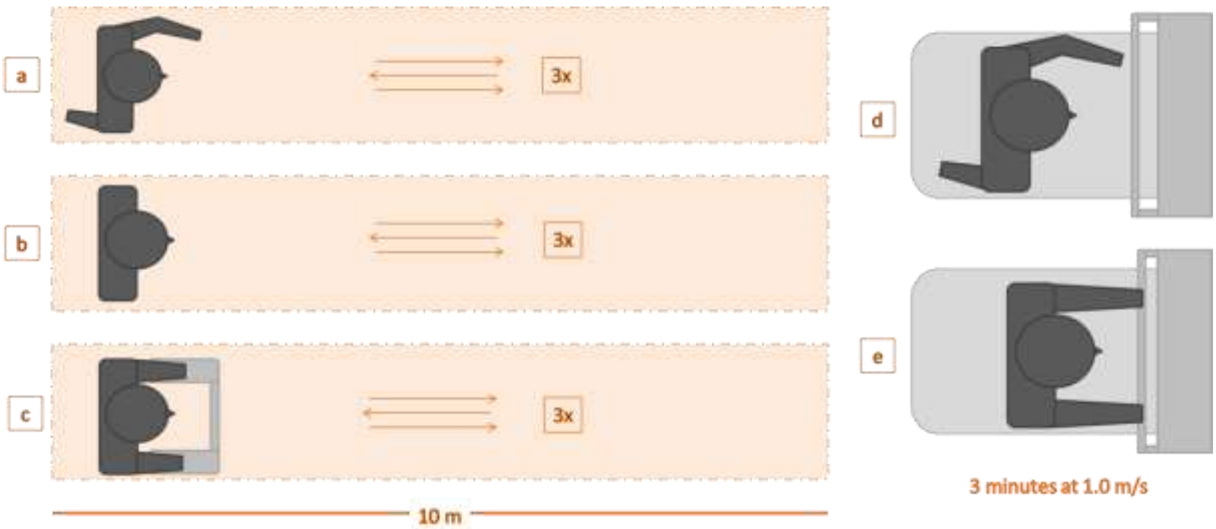

Figure 6. Five different modalities of gait performed during the experiments. (a) Gait with arm swing on ground (GAS); (b) Gait without arm swing on ground (GNAS); (c) Assisted gait (AG); (d) Gait with arm swing on treadmill (TAS); (e) Gait without arm swing on treadmill (TNAS).

3.3.4. Data analysis

Once collected to the computer, the signals were analyzed to identify the gait phases and muscle activity. From the accelerometer signals, the analysis was done following the method of Han et al. (2009), in which the gait cycle begins with the heel strike and ends in the next heel strike of the same foot, corresponding to 100% of the gait cycle. The vector module of the x and y axis of accelerometer was calculated and used to divide the gait cycle in two phases: stance, when the foot is touching the ground and sustains the body weight, and swing, when there is no foot support and the limb advances (Figure 7).

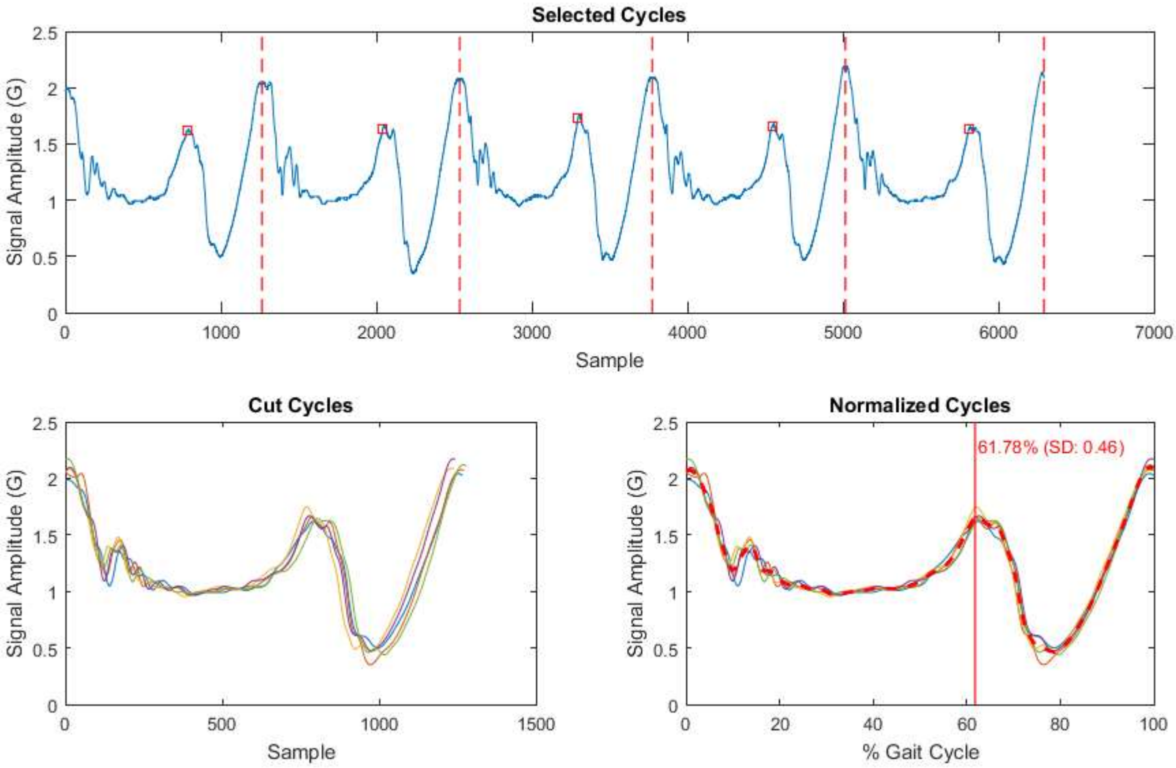

Figure 7. Accelerometer data from the volunteer V16 during the gait with arm swing on ground (GAS). Of the 10 meters walked, five gait cycles were selected and the toe-off was identified (upper graphic). After that, the cycles were cut at the peaks representing the initial contact of the right foot (left bottom) and, finally, the cycles were normalized as a percentage of gait cycle and the mean toe-off was calculated (61.78%, SD: 0.46), separating the stance and swing phases (right bottom).

The raw sEMG (Figure 8) signals were filtered using a fourth order Butterworth band-pass filter, which allows frequencies from 10 to 450 Hz to pass and have undergone full-wave rectification. After that, the root mean square (RMS) technique was applied to the enveloped signals, and was normalized by the reference method of signal maximum peak (OLSON, 2010; VAN KAMMEN et al., 2017).

The k-means clustering technique was used to allow dichotomization of the myoelectric signal in “muscle active” and “muscle inactive”. In this case, we used k = 5, and only the group with lower amplitudes was considered as “muscle inactive”, which means higher values indicate muscle activation period (DEN OTTER et al., 2007). Computing all cycles performed by each volunteer, the average activation pattern for the analyzed muscle is obtained.

For statistical analysis, the one-way ANOVA with post-hoc Tukey HSD (Honestly Significant Difference) was applied to compare the different modalities of gait and to verify if there was a significant difference between them. When p-value < 0.05, the null hypothesis was rejected.

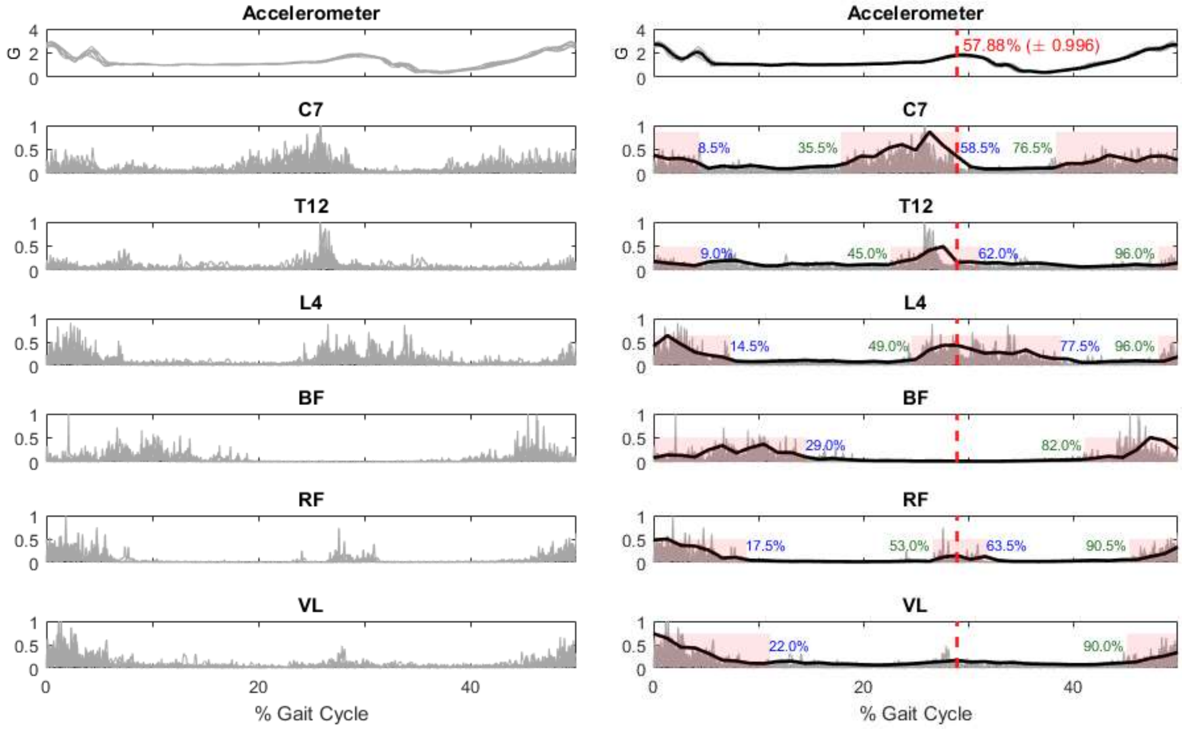

Figure 8. Muscle activation pattern of the volunteer V25 during gait with arm swing on ground (GAS). The figure on the left shows the muscle pattern obtained after full-wave rectification, filtering, normalization by the initial contact of the right foot (by the accelerometer signal) and by the method of signal maximum peak (amplitude of the EMG signal, where all are amplitude from 0 to 1). The figure on the right shows an envelope obtained by root mean square (RMS) technique and the k-means clustering technique, to identify onset and offset of muscles.

3.4. RESULTS AND DISCUSSION

Thirty healthy subjects participated of this study, whose characteristics are presented in Table 1. When asked to walk on the ground, the subjects could choose the most comfortable speed for them. Table 2 shows the average speed of each gait.

Table 1. Healthy volunteers' characteristics.

Volunteer	Gender	Age (years)	Height (cm)	Weight (kg)	BMI (kg/m^2)
V1	M	26	187	85	24
V2	M	31	178	95	30
V3	F	22	170	65	23
V4	M	25	172	65	23
V5	M	30	163	66	25
V6	M	37	184	73	22
V7	M	26	171	90	31
V8	F	33	160	56	22
V9	F	31	158	46	18
V10	F	25	163	57	22
V11	M	26	183	70	21
V12	F	29	156	58	24
V13	F	34	150	46	20
V14	M	27	170	73	25
V15	M	35	170	70	24
V16	M	29	178	62	20
V17	F	30	157	58	24
V18	F	27	160	73	29
V19	F	18	163	58	22
V20	F	22	160	57	22
V21	M	25	174	64	21
V22	M	23	178	80	25
V23	M	25	168	67	24
V24	M	25	175	79	26
V25	M	38	185	110	32
V26	F	18	156	45	19
V27	F	33	177	75	24
V28	F	24	164	65	24
V29	F	21	172	58	20
V30	F	23	170	50	17
Mean	15M/15F	27 ± 5	169 ± 10	67 ± 15	23 ± 4

M: male; F: female; BMI: Body Mass Index

Comparing the speed of the group in three different walks on the ground, we identified significant statistic difference between GAS x AG, and GNAS x AG. Free gait speeds (from GAS and GNAS evaluations) were higher than assisted gait (AG) considering the average result of the group. A reduction in the speed during the walker-assisted gait was expected, according to the literature (MARTINS et al., 2012). This reduction did not affect the phases duration because the AG toe-off value remained similar in the other gaits.

Table 2. Speed and percentage of stance phase (toe-off) in the group and statistic comparison among different gaits, showing the p-value.

		GAS	GNAS	AG	TAS	TNAS
Speed (m/s)		0.99 ± 0.11	1.03 ± 0.12	0.88 ± 0.12	1.0 (fixed)	1.0 (fixed)
Comparison – Speed (p-value)	GAS	-	0.372	**0.002**	-	-
	GNAS	0.372	-	**< 0.001**	-	-
	AG	**0.002**	**< 0.001**	-	-	-
Toe-off (%)		61.56 ± 2.76	61.48 ± 2.49	61.43 ± 2.92	62.01 ± 2.66	62.37 ± 3.82
Comparison – Toe-off (p-value)	GAS	-	0.999	0.993	0.975	0.826
	GNAS	0.999	-	0.943	0.953	0.762
	AG	0.993	0.943	-	0.939	0.740
	TAS	0.975	0.953	0.939	-	0.989
	TNAS	0.826	0.762	0.740	0.989	-

GAS: Gait with arm swing on ground; GNAS: Gait without arm swing on ground; AG: Assisted gait; TAS: Gait with arm swing on treadmill; TNAS: Gait without arm swing on treadmill.
Data in which significant statistic differences (p-value < 0.05 – one-way ANOVA with post-hoc Tukey HSD) were found are highlighted in bold.

Gait cycle begins, by convention, when there is foot contact on the ground, and ends with the next contact of the same foot. It consists of two distinct phases, called “stance phase” and “swing phase”. In our experiments, we used an accelerometer on the ankle (above the lateral malleolus) to identify the initial and final contact (toe-off) of the foot on the ground, and, consequently, obtain stance and swing phases of the gait cycle (HAN et al., 2009; LEE et al., 2010). Such as aforementioned, we analyzed five different modalities of gait with the group of volunteers, finding that the duration of the stance phase (it ends in toe-off) did not have significate differences in any of them, as can be seen in Table 2.

3.4.1. Muscle pattern

Trunk muscles play a role of preserving the stability during walking, and some authors (CECCATO et al., 2009; KARTHIKBABU et al., 2012; PEREIRA et al., 2011) suggest these muscles are activated during the gait also as anticipatory action. In Figure 9 the muscle pattern for the group, in five different gaits, is presented.

In this research, we identified two periods of activation of ES, starting shortly before the double support phases. The first period begins (1^{st} onset) at mid stance or pre-swing (stance phase) and ends (1^{st} offset) at initial swing (swing phase). The second period begins (2^{nd} onset) at mid or terminal swing (swing phase) and ends (2^{nd} offset) at the mid stance (stance phase).

Studies (CECCATO et al., 2009; KARTHIKBABU et al., 2012) show ES acts synchronously, presenting a rostrocaudal sequential activation during walking. Here, we analyzed ES muscle in three levels (C7, T12, L4), and we observed the upper level (C7) was activated previously (1^{st} onset 36.00% ± 3.09 and 2^{nd} onset 81.95 % ± 3.10 of the gait cycle in GAS), followed by T12 (1^{st} onset: 48.40% ± 2.84; 2^{nd} onset: 93.38% ± 2.91 in GAS) and lower level L4 (1^{st} onset: 51.65% ± 2.47; 2^{nd} onset: 96.25% ± 2.93 in GAS), respectively.

Ceccato et al. (2009) analyzed ES muscles bilaterally in the following levels: C7, T3, T7, T12 and L3. They observed two activation peaks during gait, one in the first double support and other in the second double support. However, our results show the activation peaks are in the two phases of double support with greater amplitude in the initial contact phase of the contralateral limb (~ 50-60% of the gait cycle). One of the possibilities is that ES muscles contract to balance the trunk and pelvis for the swing phase. In addition, these muscles may be responsible for weight transfer between limbs and thorax, and pelvis rotation inversion, such as observed by White and McNair (2002). Perry and Burnfield (2010) suggest higher sEMG amplitude of contralateral ES is due to the pelvic decline, and its ipsilateral activity decelerates the trunk progression.

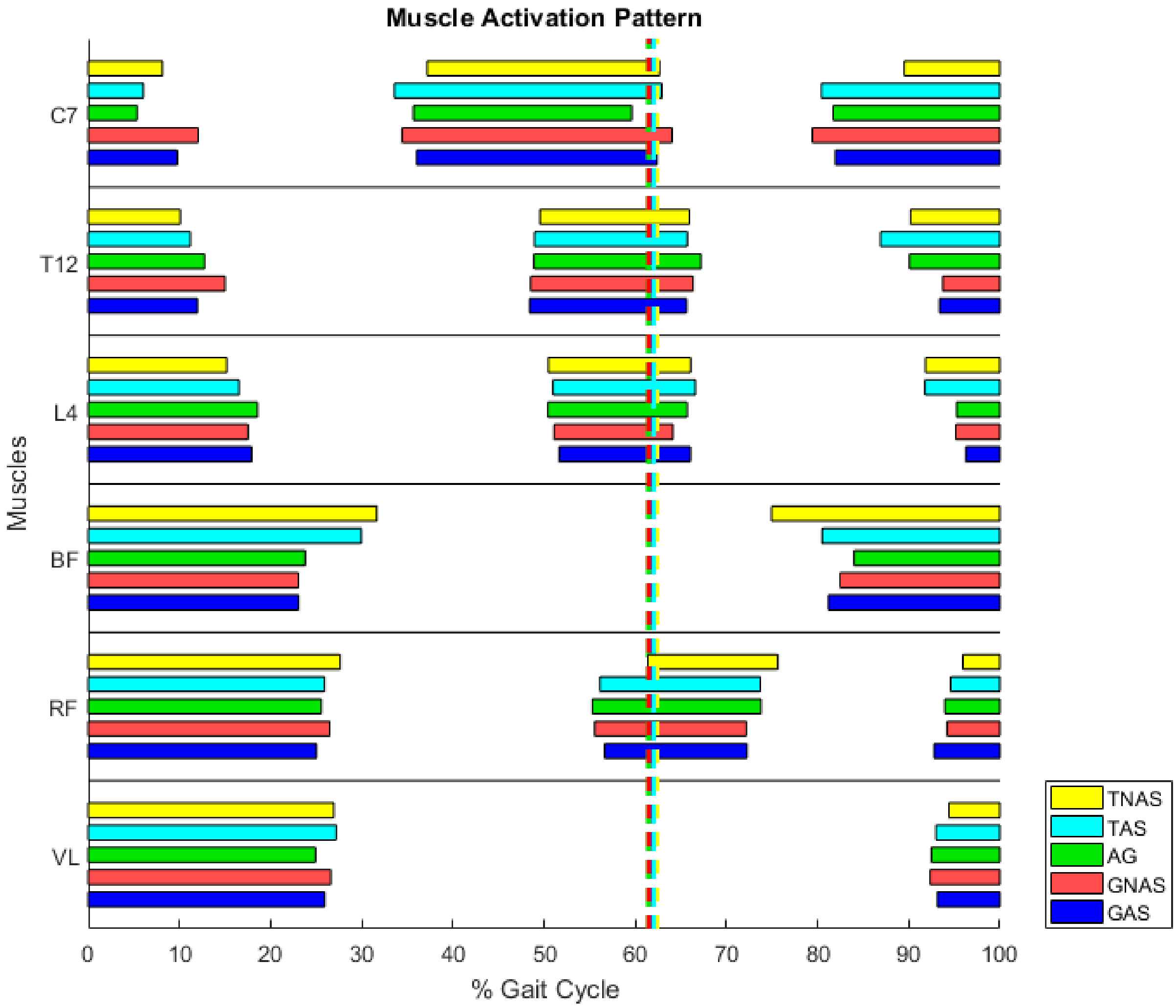

Figure 9. Mean of muscle activation pattern from 30 subjects of the control group. GAS: Gait with arm swing on ground; GNAS: Gait without arm swing on ground; AG: Assisted gait; TAS: Gait with arm swing on treadmill; TNAS: Gait without arm swing on treadmill; C7, T12 and L4 are the erector spinae levels analyzed; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis.

Comparing all muscles onset/offset (Figure 10), in GAS x GNAS and GAS x AG, we did not find significant statistic differences, which indicate the gait without arms swing (GNAS and AG) does not affect directly the normal muscle pattern during walking on ground. Likewise, TAS and TNAS only showed statistically significant difference in C7 2nd onset, indicating that ES does not present important differences with the arm swing, mainly in lower levels (T12 and L4).

As for ES muscle analysis, we observed significant statistic differences in later C7 2nd onset during TNAS when compared with all other gaits. The later C7 offset during GNAS had significant statistic difference compared to AG and TAS (2nd offset). In the 2nd activation period of T12 there were significant statistic differences: earlier onset of AG (comparing with GNAS) and TAS (comparing with free gait on ground), and earlier offset of treadmill walking comparing with free gait on ground. Finally, there were significant statistic differences in L4 2nd onset in all comparisons between ground and treadmill gaits, as treadmill gaits showed earlier onset in the swing phase. Walking on treadmill led more changes in the onset/offset muscles, already mentioned in previous studies, which occur in stride length, joint range of motion and EMG activation (LEE; HIDLER, 2008; SLOOT; VAN DER KROGT; HARLAAR, 2014). The fixed speed, possible difficulty in maintaining the balance, and not being familiar with walking on the treadmill can be the main cause of the observed changes.

The analysis of the lower limb muscle pattern is well described in the literature. RF, BF and VL muscles show clear onset/offset according to (CRIEKINGE et al., 2018; PERRY; BURNFIELD, 2010; WARD et al., 2018). Regarding activation, in our study the BF muscle was the most affected by the treadmill, being active for longer period. The BF onset during TNAS was the earliest and the BF offset during treadmill occurred later than gait on the ground. RF muscle was activated later in TNAS, presenting significant statistic differences in 1st onset (comparing with all gaits), 1st offset (GAS and GNAS) and 2nd onset (GAS). VL onset/offset did not present significant statistic differences.

Gait patterns of males and females differ mainly due to the anthropometric differences. Anders et al. (2009) found ES exhibits higher amplitudes during the contralateral initial contact in females, occurring in the sagittal plane. In our study, the comparison of speed, toe-off and muscle onset/offset of males and females within each gait did not present significant statistic differences, considering the p-value (for GAS the p-value = 0.110; GNAS p = 0.097; AG p = 0.354; TAS p = 0.332 and; TNAS p = 0.397).

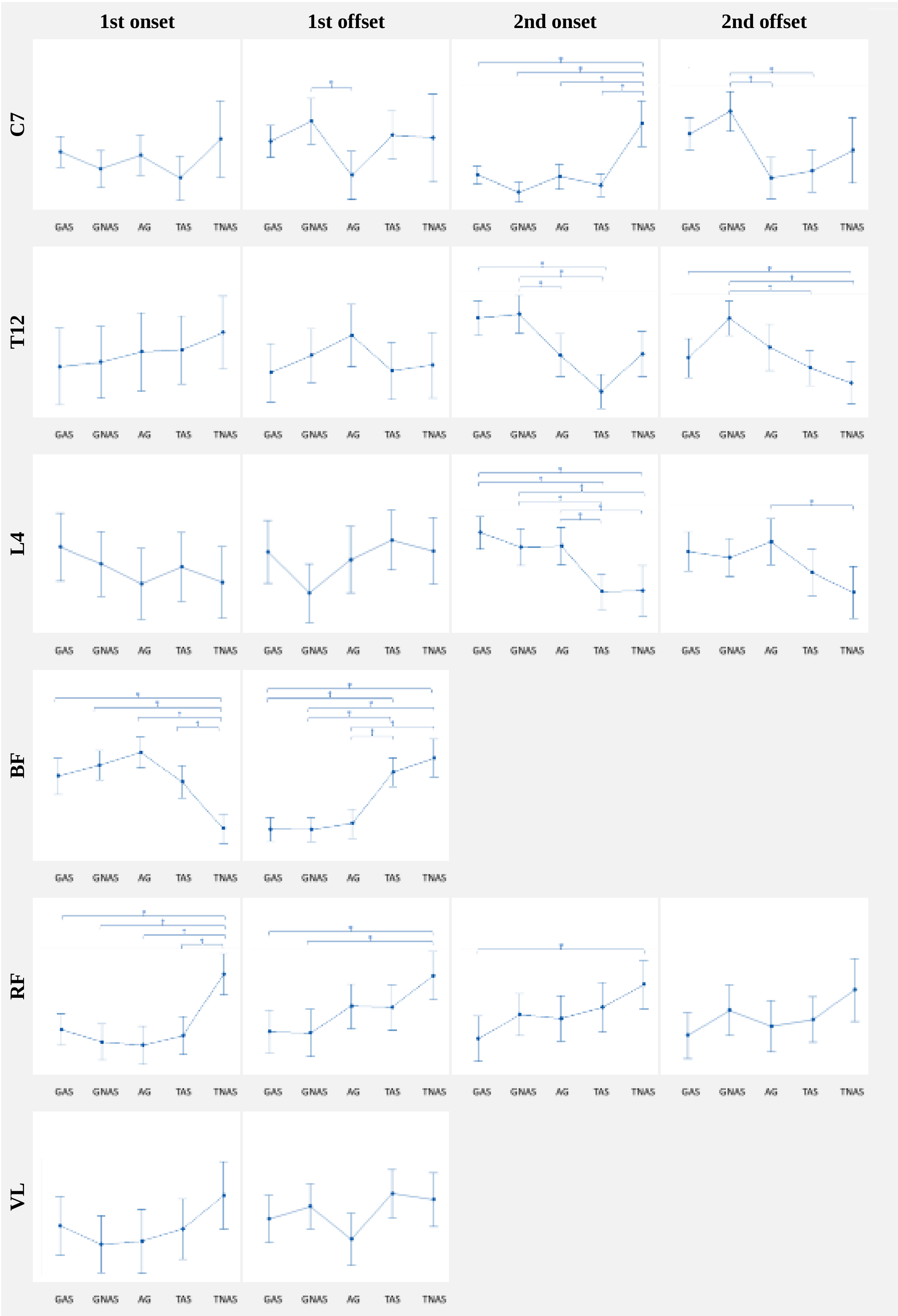

Figure 10. Statistic comparison of muscles onset and offset among different gaits, using one-way ANOVA with post-hoc Tukey HSD. The symbol (*) indicates there is significant statistic difference, with p-value < 0.05. GAS: Gait with arm swing on ground; GNAS: Gait without arm swing on ground; AG: Assisted gait; TAS: Gait with arm swing on treadmill; TNAS: Gait without arm swing on treadmill; C7, T12 and L4 are the erector spinae levels analyzed; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis.

3.5. CONCLUSIONS

Trunk muscles have an important function in motor tasks, acting directly in the balance and posture during gait. Five modalities of gait were compared in this investigation: gait with and without arm swing on the ground, assisted gait and gait with and without arm swing on a treadmill. The assisted gait had lower speed when compared with gait on the ground. All the toe-off identified during gaits had similar value.

The data in this study indicates ES muscle has a rostrocaudal sequential activation pattern in the gait cycle, being two periods of activation predominantly in the double support phases. Our findings showed gait without arms swing does not affect the ES muscle pattern during walking.

Significant statistic differences were found in: later C7 2nd onset during TNAS; later C7 offset during GNAS (comparing with AG and TAS); earlier T12 second onset of AG (comparing with GNAS) and TAS (comparing with free gait on ground), and earlier T12 second offset of treadmill comparing with free gait on ground; L4 2nd onset in treadmill showed earlier onset in the swing phase. Walking on treadmill had a greater influence the on onset/offset muscles. Finally, the comparison among the onset/offset of males and females within each of the different gait did not present significant statistic differences.

As a contribution of this study, these results can be used to compare with pathological gait, verifying progression of rehabilitation of people with mobility disorders or for diagnosis of diseases that affect gait and balance, as well as for studies about posture of people in gait assisted walkers.

4. IDENTIFICATION OF NEUROMUSCULAR FATIGUE DURING GAIT ON TREADMILL AND ISOMETRIC EXERCISES THROUGH SHORT-TIME FAST FOURIER TRANSFORM

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4.1. ABSTRACT

Neuromuscular fatigue is a strength reduction generated by the muscle, which can be detect through the decrease of the median frequency (MDF) of electromyographic signals. The objective of this study is to identify neuromuscular fatigue during isometric exercises and walking on a treadmill, using the Short-Time Fast Fourier Transform (STFFT). Ten healthy participants performed three isometric exercises until the task failure (with three lower-limb muscles analyzed), and walked on a treadmill for 3 minutes at 1.0 m/s (with three lower-limb muscles and a trunk muscle — in three levels — analyzed). During the isometric tasks, there were significant decreases in the tibialis anterior and vastus lateralis, with reduction of the MDF (96.2 ± 4.7% and 95.4 ± 3.4%, respectively). Only the L4 level of the erector spinae presented a significant slope regression (-0.050 ± 0.028 Hz/s). All the lower-limb muscles also showed reduction in their MDF. The STFFT seems to be useful to detect MDF changes during non-strenuous exercise, which can be used to analyze the natural development of neuromuscular fatigue in non-strenuous tasks in people with predisposition to fatigue.

Keywords: Neuromuscular Fatigue; Median Frequency; sEMG; Isometric Exercises; Gait.

4.2. INTRODUCTION

Neuromuscular fatigue can be defined, biomechanically, as a decrease in the ability of the muscle to generate force or maintain a specified force output, induced by exercises, but reversible after rest (CHANG et al., 2017; OLSON, 2010; TWOMEY et al., 2017). It may result in altered muscle patterns and strength, and affect motor control and coordination (CHANG et al., 2017). At the area of neurophysiology, neuromuscular fatigue is considered as a decrease in median frequency output of electromyographic signals over time (MINNING et al., 2007; OLSON, 2010). Several activities in professional tasks may be affected by neuromuscular fatigue and, therefore, several studies have been carried out about this subject in, for instance, pilots (HONN et al., 2016), fire-fighters (DAWSON et al., 2015), and military personnel (QU; YEO, 2011). In addition, its presence is often in some diseases, as multiple sclerosis (KRUPP; SERAFIN; CHRISTODOULOU, 2010) and stroke (ANGELOVA et al., 2018; GERRITS et al., 2009; XU; CHU; ROGERS, 2014), which may interfere in the rehabilitation process, reducing the quality of life of the affected.

Researches about neuromuscular fatigue, normally induced it by strenuous exercises, until the subject either reports exhaustion (VIEIRA et al., 2016), fails the task or reaches a predetermined time (KENNEDY et al., 2011; KUTHE; UDDANWADIKER; RAMTEKE, 2018; MINNING et al., 2007; TWOMEY et al., 2017). Electrical stimulation is another way to induce fatigue (GERRITS et al., 2009).

Surface Electromyography (sEMG) is widely used to examine the muscle function, including fatigue (KUTHE; UDDANWADIKER; RAMTEKE, 2018), as there are changes in both time and frequency domains as fatigue develops (ASEFI et al., 2016). In the time-domain, mean absolute value, root mean square, and zero crossing per second of the sEMG signal can be used to detect neuromuscular fatigue (KUTHE; UDDANWADIKER; RAMTEKE, 2018), however these features are not as effective as the frequency-domain features (THONGPANJA et al., 2013). Mean frequency (MNF) and median frequency (MDF) are often used in frequency-domain (AL-MULLA; SEPULVEDA; COLLEY, 2011), which have a shifting to lower frequencies when the fatigue occurs (KUTHE; UDDANWADIKER; RAMTEKE, 2018).

Two exercises modalities are often investigated to study the occurrence of fatigue, being them isometric contractions, in which there is no alterations in muscle length, e.g., the subject remains in a static position (KUTHE; UDDANWADIKER; RAMTEKE, 2018); and dynamic contractions, in which there are muscle movements, for instance, during cyclic activities, as gait and pedaling. Such as aforementioned, changes in MNF and MDF are tracked over time in both exercises modalities, when their changes are tracking over time (ASEFI et al., 2016; THONGPANJA et al., 2013). Protocols designed for detecting neuromuscular fatigue in isometric exercises normally use the maximal voluntary contraction (MCV), measuring how the behavior of MCV changes before and after a fatiguing task (TWOMEY et al., 2017). For dynamic exercises, the same idea can be applied.

Many studies have also investigated the neuromuscular fatigue during walking, for instance, Qu and Yeo (2011), Qu (2015), Hatton et al. (2013) and Vieira et al. (2016) evaluated the fatigue in different modalities of gait, using kinematic parameters before and after inducing fatigue. On the other hand, Janssen et al. (2011) analyzed gait patterns through ground reaction forces before, during and after a fatigue protocol.

Chang et al. (2017) analyzed the influence of fatigue in muscle patterns from 25 healthy individuals, who walked on a treadmill at 1.3 m/s for 20 s before and immediately after they were submitted to an exercise protocol (series of walking, jump squats and lateral hops). In other research, Barbieri et al. (2013) analyzed the effects of fatigue on the kinematic and kinetic in adults during gait (path of 8 m at self-selected speed) before and after a fatigue protocol, in which the fatigue was induced in quadriceps muscles using sit-to-stand task until the failure task, loss of speed or after 30 min. In both studies, changes in frequency were not analyzed.

Some researches have applied the short-time Fast Fourier Transform (STFFT) technique to determine the MNF and/or MDF of the power spectra of the sEMG, analyzing different window lengths. For instance, 100 ms, in static sub-maximal trunk extension (OLSON, 2010), 256 ms, in back and hip muscles during an isometric fatiguing test (COOREVITS et al., 2008), 512 ms, in isometric contractions of the biceps brachii (KUTHE; UDDANWADIKER; RAMTEKE, 2018), and 341 ms in elbow flexion (ANGELOVA et al., 2018). Hollman et al. (2013), during a fatigue test in which sEMG signals from the gluteus maximus and semitendinosus were analyzed, and using five different window lengths, found that window lengths do influence the MDF variability, however, they verified that the MDF slopes are equivalent across all conditions.

Such as aforementioned, studies have been conducted to detect neuromuscular fatigue during gait, inducing fatigue through strenuous exercises, and, finally, evaluating the influence of fatigue on the gait with the already fatigued muscle. However, neuromuscular fatigue is a continuous process, which develops itself gradually during the physical activity, and not in a specific moment, such as the task failure (TWOMEY et al., 2017). Therefore, the objective of this study is to identify the neuromuscular fatigue in lower-limb muscles during isometric exercises and in lower-limb and trunk muscles walking on treadmill at normal speed, using the STFFT. Our hypothesis is that the decrease in MDF is detectable during the fatigue process in non-strenuous exercises.

4.3. MATERIAL AND METHODS

4.3.1. Volunteers

Ten individuals, five of them males and five females, aged 21 to 38 years, and without motor impairment participated of the experiments in the laboratory of the Assistive Technology Group at the Federal University of Espirito Santo (UFES). The study had approval of the UFES's Ethic Committee, and all volunteers signed the Free and Informed Consent Form.

In order to participate in the experiments, the subject should have the following characteristics: neither suffering from motor impairment, both musculoskeletal and joint nor pains at lower limbs, upper limbs and trunk, sufficient cognitive, visual and language skills to understand and follow the test instructions. Exclusion criteria include the presence of cardiorespiratory or other diseases that interfere with gait and have performed strenuous physical exercises 24 hours before the test.

4.3.2. Data Acquisition

The acquisition of the sEMG signals was done using the equipment EMG System do Brasil Ltda®, which has 16-bit analog/digital conversion resolution, amplifier gain up to 2000 V/V, common mode rejection > 100 dB, input impedance of 10^9 Ω, and sampling frequency of 1000 Hz.

The position of the electrodes was determined following the recommendations of the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM, 2016), and such as suggested by (DE SÈZE et al., 2008; SWINNEN et al., 2012). In both stages, a reference electrode was placed on the lateral malleolus, and the dominant limb was analyzed. In the first stage of these experiments, the following muscles were analyzed: tibialis anterior (TA), gastrocnemius medial (GM), and vastus lateralis (VL). In the second one, the muscles analyzed were erector spinae (ES) in the C7, T12 and L4 levels, biceps femoris (BF), rectus femoris (RF), and vastus lateralis (VL).

4.3.3. Experimental Protocol

The experiments were performed in two stages. Initially, the participants performed isometric contractions, in order to verify changes in the median frequency (MDF) during the completely static exercise. As second stage, the data obtained during gait on treadmill were analyzed using the same processing of the first stage.

As these exercises were performed without the addition of external weights, they did not demand maximum muscle strength and are, therefore, called submaximal static exercises (OLSON, 2010). In these cases, it is expected an increased recruitment of motor units, with high signal amplitude. Moreover, it is expected a decreasing in MDF over time, mainly due to the decrease of the conduction velocity of the motor action potentials on the muscle membrane (KONRAD, 2005).

Before starting the experiments, the International Physical Activity Questionnaire - Short Form (IPAQ-SF) (ANNEX C) was applied in order to assess the level of physical activity of each individual, and classify them as physically active or sedentary. This classification was obtained by calculating the time and intensity of physical activity practice the volunteer has made for the week before the test. In the IPAQ-SF, the individual can be classified into four categories: very active, active, irregularly active A, irregularly active B, and inactive (CRAIG et al., 2003; MATSUDO et al., 2001).

4.3.3.1. First stage – Isometric Exercise

In the first stage of the neuromuscular fatigue tests, the volunteers performed the following tasks (Figure 11):

- The subject remained seated with the heel on the ground and the tip of the foot elevated at maximum angulation, maintaining this position until exhaustion of the tibialis anterior (TA) muscle;
- The subject remained in the toe-lift position for isometric contraction of the gastrocnemius medialis (GM) muscle, and the position was maintained until task failure, i. e., until he/she no longer supported the predetermined position;
- Finally, the volunteer kept the task in an isometric contraction of the vastus lateralis (VL) muscle, in the final position of the traditional squat with the knees at 90º, but leaning against the wall, and remained thus until the isometric task failure.

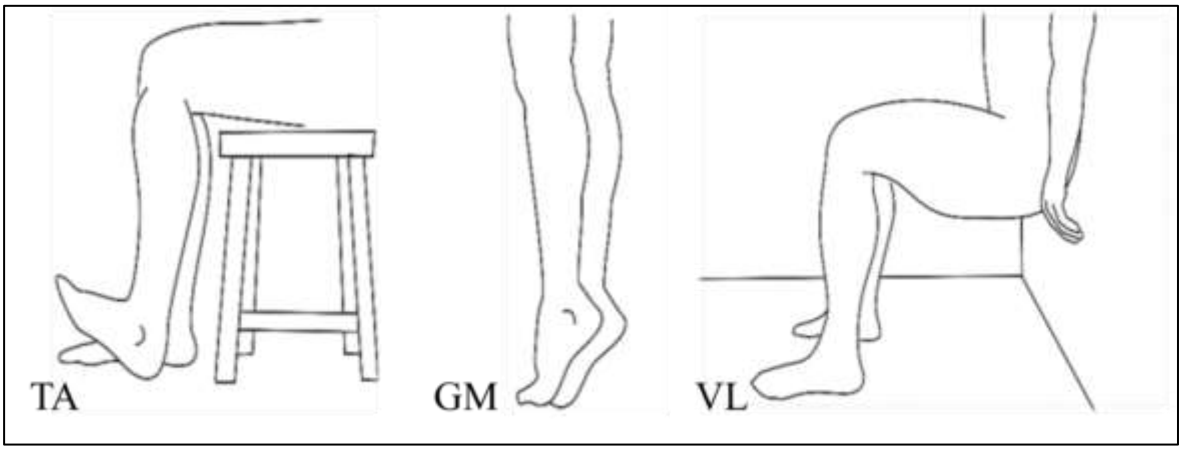

Figure 11. Positions maintained for the isometric contraction of the tibialis anterior TA (left), gastrocnemius medialis GM (middle) and vastus lateralis VL (right) muscles.

4.3.3.2. Second stage – Gait on the treadmill

After an interval of at least a week, the participant returned to the laboratory to perform the second stage of the experiment, which was walk on a treadmill at fixed speed of 1.0 m/s for 3 minutes with natural arm swing.

4.3.4. Analysis of the sEMG Signals

In both stages of the experiments, initially the sEMG signal was filtered using a fourth order Butterworth bandpass filter of 10-450 Hz.

Such as aforementioned, the MNF and MDF are considered the gold standard for neuromuscular fatigue analysis, however, MDF is less affected by random noise and more influenced by neuromuscular fatigue than MNF (PHINYOMARK; PHUKPATTARANONT; LIMSAKUL, 2012). Therefore, we used only the MDF of the power spectra, which was determined by STFFT using a window length of 256 ms, such as done by (COOREVITS et al., 2008).

MDF is defined as the frequency in which the sEMG power spectrum is divided into two regions with equal value (KUTHE; UDDANWADIKER; RAMTEKE, 2018). As fatigue accumulates, that value decreases, therefore, it is expected that the MDF of sEMG reduces after gait (KIM et al., 2013).

A regression slope of MDF over time towards lower frequencies can be used as a fatigue index for the investigated muscle, in which as larger the negative slope value as greater the neuromuscular fatigue (KONRAD, 2005; MINNING et al., 2007). The following linear regression function can relate MDF and time during the muscle activity:

$$y = mx + c \tag{1}$$

where y is the MDF, x is time interval, m is the regression slope, and c is the bias (KUTHE; UDDANWADIKER; RAMTEKE, 2018).

Thus, for calculating the percentage of MDF decrease or increase, a linear regression of the signal was used, considering the first point of the line as reference (100%), with the last point calculated, in percentage, with respect to the initial point. The signal processing was performed in MatLab (2016a), using custom algorithms.

4.3.5. Statistics

The normality data were tested through Shapiro-Wilk test, and did not present a normal distribution. Thus, the average first point was compared with the average last point of the regression slope using the Wilcoxon signed-rank test. Additionally, the coefficient of variation (CV) was used to verify how much the values varied within a sample, in which CV lower than 15% considered low dispersion, and higher than 30%, high dispersion of the values.

4.4. RESULTS AND DISCUSSION

4.4.1. First stage – Isometric Exercises

In the first stage, the participants (Table 3) were asked to perform three different isometric exercises, each one designed specifically to activate the muscle to be studied. Figure 12 shows the linear regression of the MDF over time during each isometric exercise, and Figure 13 presents how much the decrease or increase of MDF was in these tasks. The values of each individual and the group mean are shown in percentage. In both figures, the three muscles are represented by different graphics.

Table 3. Characteristics of participants and their IPAQ-SF.

	Information before test			
	Age	Gender	IPAQ-SF	Dominant side
V1	29	M	IA-A	R
V2	21	F	Active	R
V3	22	M	Active	R
V4	35	M	IA-B	R
V5	29	F	Active	R
V6	24	F	Active	R
V7	33	M	Active	L
V8	32	F	Active	R
V9	24	M	Active	R
V10	38	F	IA-B	R

M: male; F: female; R: right; L: left.
IA-A: irregularly active A; IA-B: irregularly active B.

TA is a shank muscle, responsible, mainly, for the dorsiflexion of the ankle joint (SENIAM, 2016). During the TA task, the participant remained comfortably seated, and performed a dorsiflexion until he/she was no able to keep this position. The average duration for this exercise was 270 ± 98 s (CV = 27%), the longest exercise in this study, as it did not require a weight support, and had a moderate dispersion in duration. The slope regression obtained in this task was -0.057 ± 0.028 Hz/s (Table 4), which shows a decline in the MDF over time, such as expected. Calculating the percentage considering the last point of the regression line and as reference (100%) the first point, the percentage value found was 96.2 ± 4.7%. The comparison between the mean values of both first and last points indicated this reduction was statically significant ($p = 0.047$) (Table 4).

GM is a calf muscle that is part of the triceps surae group, together with the gastrocnemius lateralis and soleus muscles. This group is the main responsible for the plantar flexion of the ankle (SENIAM, 2016). In the GM task, the participant stood, resting on the feet tips and with his/her fingertips against the wall just to help keep the balance. The mean duration of this exercise was 253 ± 80 s (CV = 32%). In Table 4 is showed that GM had the lowest slope regression in the isometric exercises (-0.025 ± 0.019 Hz/s), and despite having MDF decreased it was not significant ($p = 0.333$). it seems that this muscle may not have reached a detectable level of fatigue, as the GM task depends on the balance, in which if the subject was not able to maintain his/her position, that was considered as the end of the exercise.

Table 4. Variation of the values obtained for the median frequency (MDF) of the sEMG signals during the first stage of the experiments.

Muscles	Slope regression (Hz/s)	% of the last point in relation to the first point	p-value (last point x first point)
TA	-0.057 ± 0.028	96.2 ± 4.7	**0.047**
GM	-0.025 ± 0.019	98.7 ± 13.8	0.333
VL	-0.053 ± 0.048	95.4 ± 3.4	**0.007**

TA: tibialis anterior; GM: gastrocnemius medialis; VL: vastus lateralis.
Data in which significant statistic differences (p-value < 0.05 – Wilcoxon's test) were found are highlighted in bold

Finally, VL muscle is part of the quadriceps group (rectus femoris, vastus lateralis, vastus intermedius and vastus medialis), located in the anterior thigh. The vastus muscles are responsible for the knee extension (SENIAM, 2016). The VL task had the shortest duration (101 ± 48 s, CV = 48%), as the body center of mass was not aligned with the legs, in which the quadriceps muscles sustained the weight of the upper body to compensate gravity. Both GM and VL tasks presented time duration with high dispersion, indicating heterogeneous levels of resistance in the group analyzed. VL muscle also presented a significant decrease of the MDF ($p = 0.007$), such as expected, and the slope regression was -0.053 ± 0.048 Hz/s (Table 4).

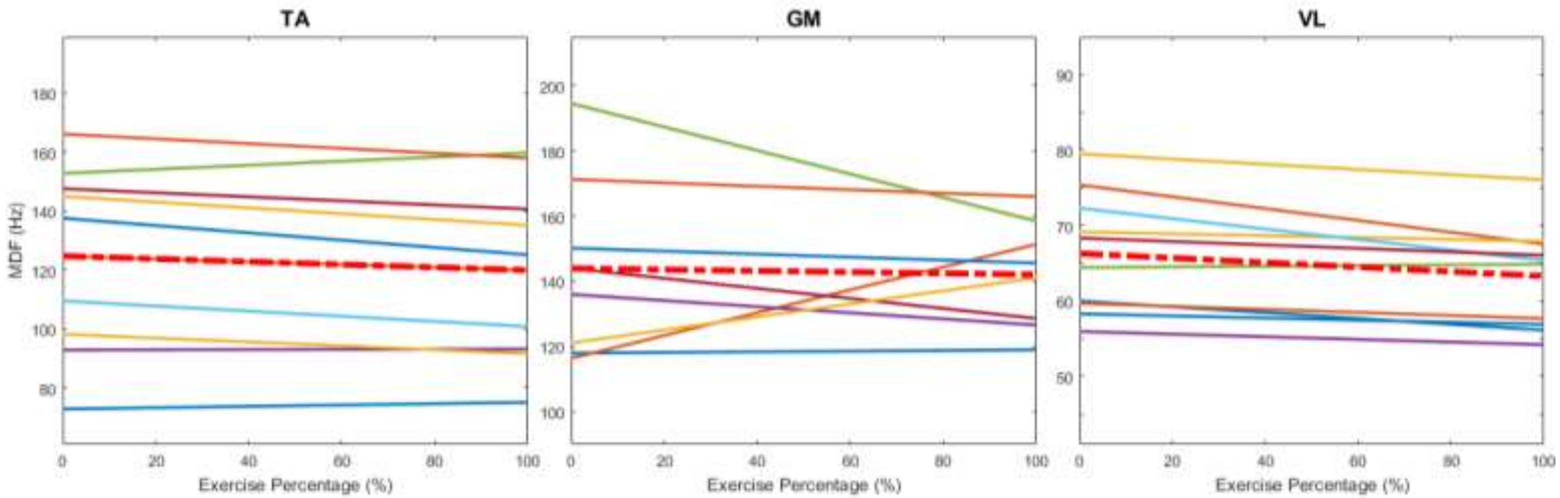

Figure 12. Regression line of the median frequency (MDF) over time during the isometric task for each muscle. The red dotted line indicates the group average of the regression lines; the others lines represent the result of each volunteer. TA: tibialis anterior; GM: gastrocnemius medialis; VL: vastus lateralis.

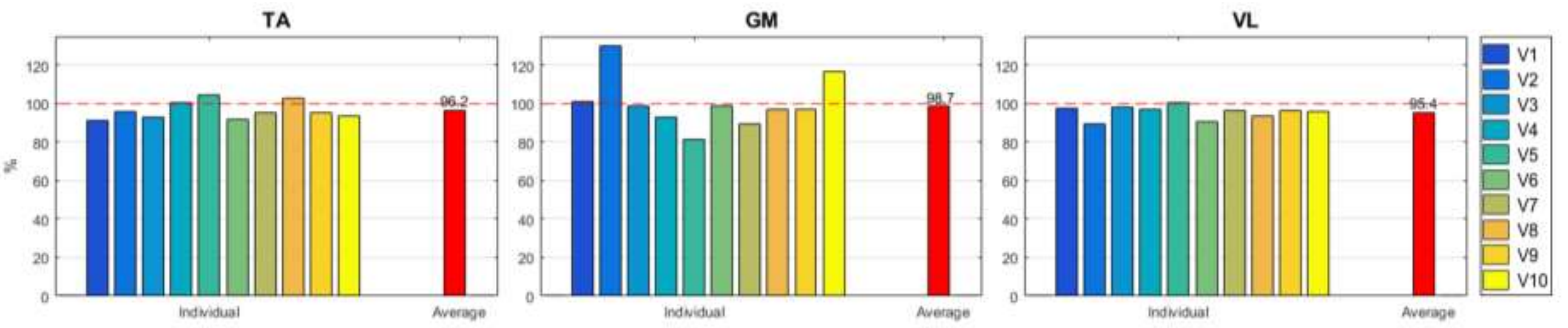

Figure 13. Percentage of decrease or increase of the final median frequency (MDF) of the isometric exercise compared to the initial MDF (considered as 100%) for each muscle. The red bar represents the group average, and the others bars represent each volunteer. TA: tibialis anterior; GM: gastrocnemius medialis; VL: vastus lateralis.

4.4.2. Second stage – Gait on treadmill

Dynamic exercises depend on the movement of many joints and muscles simultaneously. For fatigue identification, the analysis of several muscles is more stable and/or valid than the measurement obtained from a single muscle (LARIVIÈRE et al., 2002). Therefore, in this study, six muscles with functions directly associated with the walking were chosen.

Figure 14 shows the changes in MDF for the sEMG signals from six muscles (ES on levels C7, T12 and L4, BF, RF and VL) during gait performed by the volunteer 1 (V1). The equation of right side of the graphic provides the slope value, which indicates how much variation has occurred and if it is positive or negative (See Equation 1).

ES is considered the main muscle of the back, which extend by the spinal column from the skull until the pelvic region (CIONI et al., 2010; DE SÈZE et al., 2008). In addition to acting in the stability, ES is involved in the motion during gait (DE SÈZE et al., 2008; ZOFFOLI et al., 2017). Accordingly to Table 5, C7 and T12 levels of ES did not present significant changes in the MDF during the gait, while L4 had reduction of the MDF to 92.9 ± 11.5% respect to the reference. On the other hand, C7 and T12 had both lower amplitude and frequency than L4 (Figure 15) and, therefore, the measurement equipment may not have enough sensitivity and, additionally, signal noises could have interfered in the detection of little variations. Also, as the walking was at moderate speed and short duration for healthy individuals, the neuromuscular fatigue in the ES may have been developed in low level, which makes the detection difficult.

BF muscle belongs to the hamstring group (semitendinosus, semimembranosus and BF), positioned in the posterior thigh, which has as function the knee flexion (SENIAM, 2016), and acts in the propulsion phase of the gait (RAJA; NEPTUNE; KAUTZ, 2012). During the gait on treadmill, BF presented a significant change in MDF, decreasing to 93.2 ± 7.8% from the initial value (Table 5).

Table 5. Variation of the values obtained for the median frequency (MDF) of the sEMG signals during the second stage of the experiments.

Muscles	Slope regression (Hz/s)	% of the last point in relation to the first point	p-value (last point x first point)
C7	-0.001 ± 0.003	98.5 ± 3.5	0.169
T12	0.0002 ± 0.0002	100.4 ± 2.8	0.386
L4	-0.050 ± 0.028	92.9 ± 11.5	**0.029**
BF	-0.024 ± 0.015	93.2 ± 7.8	**0.028**
RF	-0.034 ± 0.004	92.6 ± 11.3	**0.022**
VL	-0.016 ± 0.009	95.3 ± 10.6	**0.048**

C7, T12 and L4 are the erector spinae levels; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis.
Data in which significant statistic differences (p-value < 0.05 – Wilcoxon's test) were found are highlighted in bold

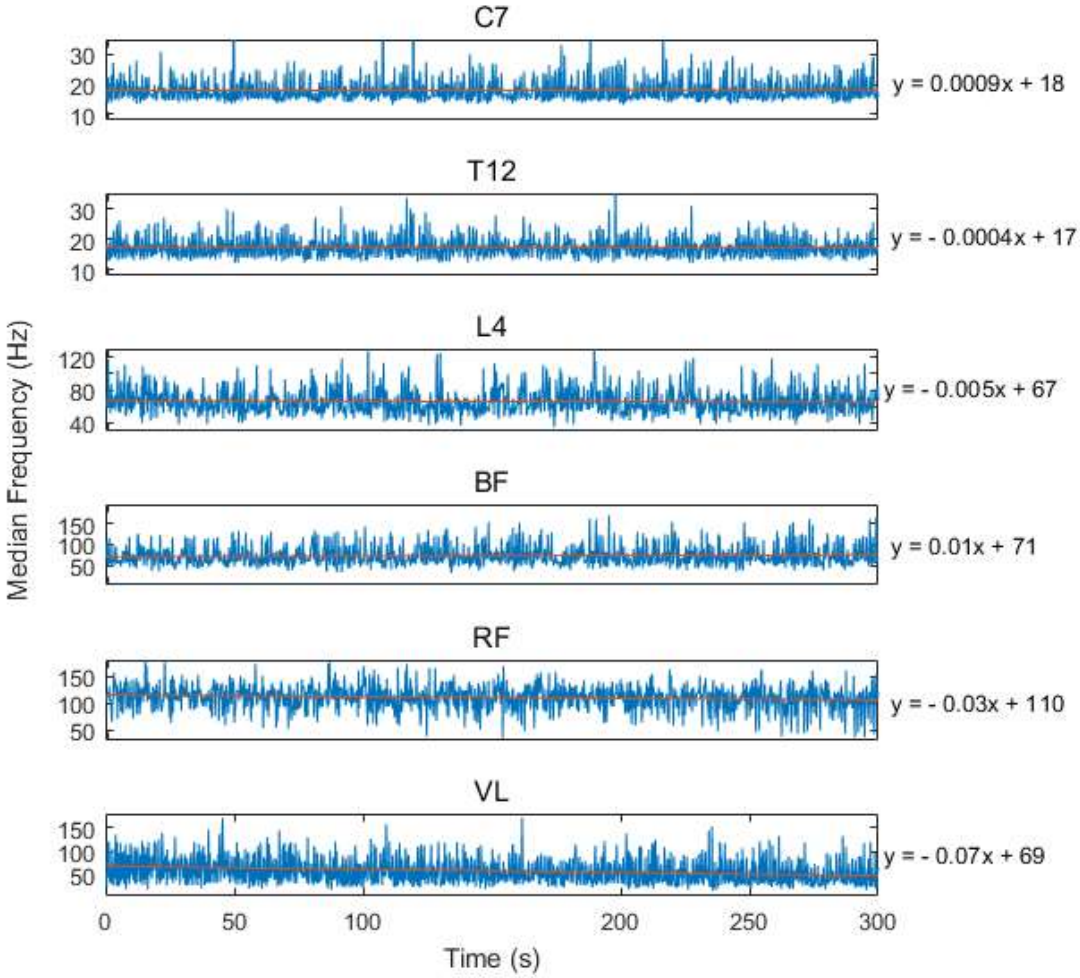

Figure 14. Changes in the median frequency (MDF) during gait for volunteer 1. The linear regression function is shown (red line), which indicates, through its slope, the behavior of the MDF during the task. The decline in the regression line indicates there is a decrease in MDF. C7, T12 and L4 are the erector spinae levels; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis.

RF and VL are quadriceps femoris muscles, which have an important role in the gait. This group counteracts the knee flexion performed by the hamstring muscles, both to decelerate the excessive flexion in the swing phase and to allow the fully knee extension in the initial contact. Additionally, RF assists the limb advancement as it is also a hip flexor (PERRY; BURNFIELD, 2010). Both muscles presented a significant negative slope (Figure 15 and Table 5), which may indicate the presence of neuromuscular fatigue during the walking. All the changes in MDF during gait are shown in Figure 16.

Through the IPAQ-SF, volunteers were classified as: 7 physically actives, 1 irregularly active A, and 2 irregularly active B (Table 3). The level of physical activities can modify the fatigue development, in addition to its effects in the motor performance, being inactive individuals more affected than the active ones (BARBIERI et al., 2013). In this study, the level of physical activity was not related to the reduction of MDF as in the isometric exercises, the decrease in the mean of three muscles was higher than 5% to V6 and V7, who were considered active subjects. In the case of walking, V6, V7, V8 and V10 presented a decrease higher than 5% in the mean values of the six muscles analyzed. It is worth to emphasize that, among those volunteers, only V10 was not physically active.

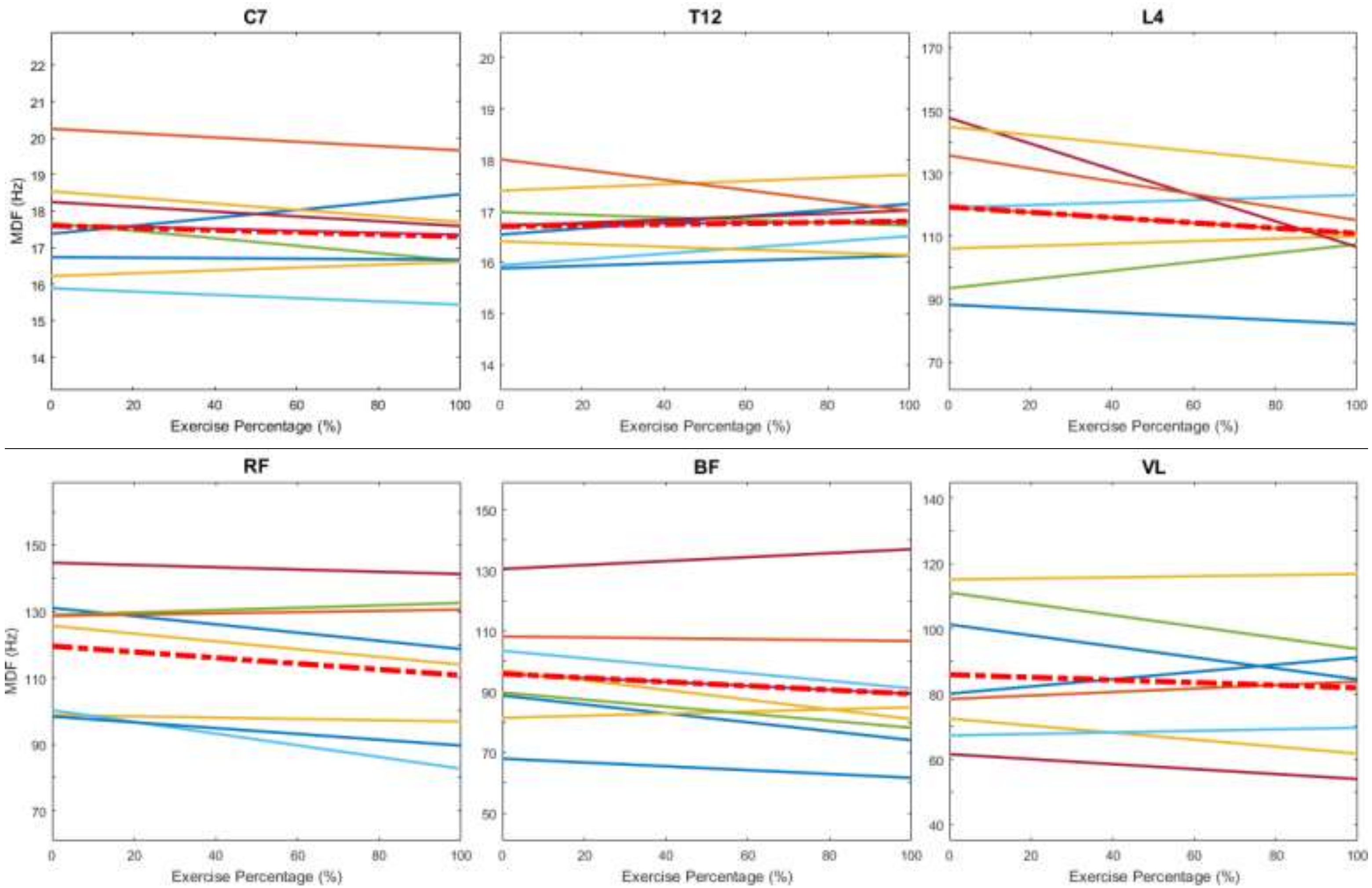

Figure 15. Regression line of the median frequency (MDF) over time during the gait for each muscle. The red dotted line indicates the group average of the regression lines, and the others lines represent the result of each volunteer. C7, T12 and L4 are the erector spinae levels; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis;

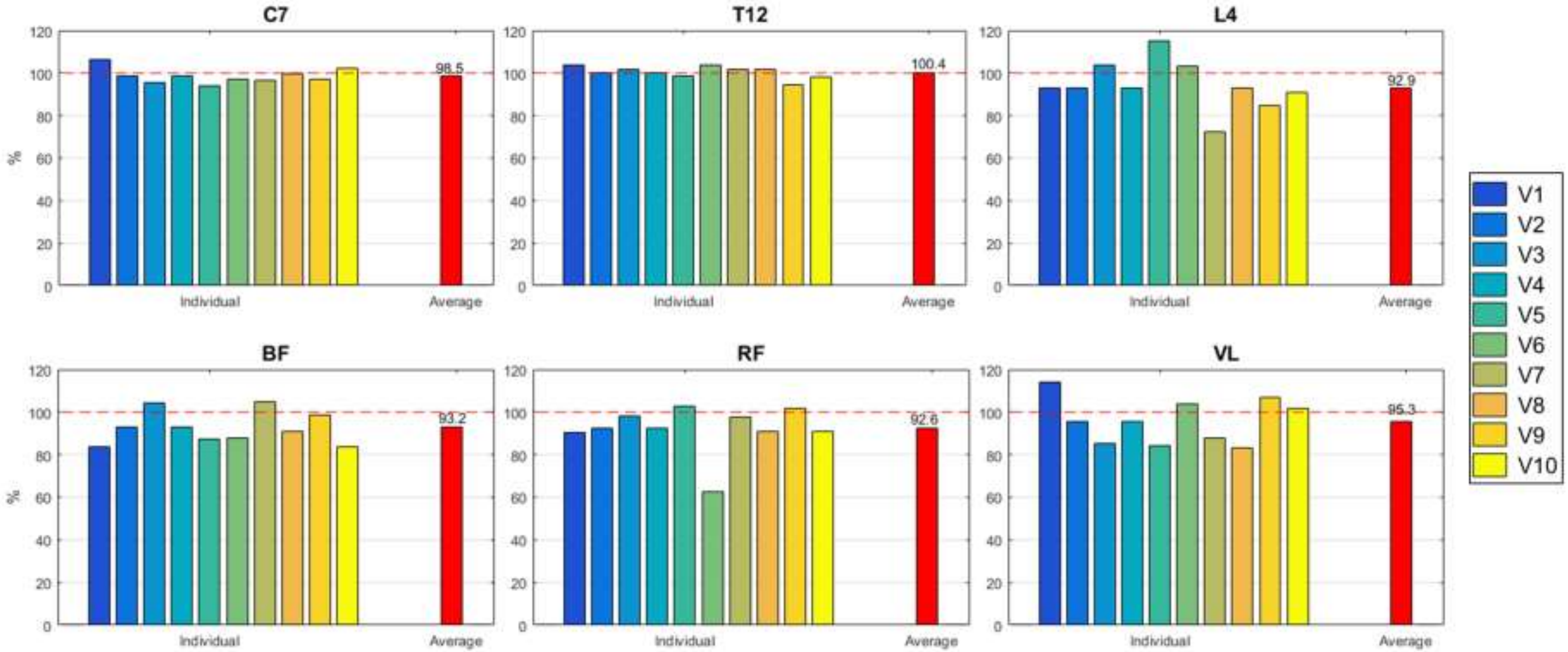

Figure 16. Percentage of decrease or increase of the final median frequency (MDF) of the exercise compared to the initial MDF (considered as 100%) for each muscle during walking. The red bar represents the group average, and the others bars represent each volunteer. C7, T12 and L4 are the erector spinae levels; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis.

4.5. CONCLUSIONS

According to literature, neuromuscular fatigue decreases the ability to perform physical activities. Using sEMG, fatigue can be identified through frequency domain analysis, in which the shift from the median frequency (MDF) to lower values indicates an increase in neuromuscular fatigue, in both isometric and dynamic contractions.

Thus, the neuromuscular fatigue may limit the exercise performance of healthy individuals in extreme or repetitive movements. However, fatigue can also occurs in simple and light tasks, as isometric exercises and gait, in people with certain diseases or disabilities, which has been little studied in the literature.

This study found that MDF of the sEMG signals shows a significant decline in TA and VL muscles during isometric exercises, and also in L4, BF, RF and VL muscles during gait. The short-time Fast Fourier Transform (STFFT) technique was used here, which has been considered useful to identify fatigue in both exercise modalities. More studies on this subject are necessary to corroborate our results, preferably with a higher sample size and discriminating groups by levels of physical activities.

5. ELECTROMYOGRAPHY ANALYSIS OF TRUNK AND LOWER-LIMB MUSCLES OF POST-STROKE INDIVIDUALS DURING FREE AND WALKER-ASSISTED GAIT

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** Submitted to Disability and Rehabilitation*

5.1. ABSTRACT

Purpose: To compare the electromyographic activity of trunk and lower limb muscles during free and walker-assisted gait in post-stroke and healthy individuals, mainly timing activation, symmetry, duration of activation in gait phase, and neuromuscular fatigue. **Methods:** Ten post-stroke and 30 healthy individuals participated of the experiments. An accelerometer was used to identify gait phases and the analyzed muscles were erector spinae (ES), biceps femoris, rectus femoris, and vastus lateralis. **Results:** In the stroke group, the ipsilateral limb had a longer stance phase than contralateral in both gaits and the walker did not modify the phases duration. ES muscle presented a sequential activation beginning on the upper level. Contralateral ES muscle of the stroke group had longer activation near the toe-off than ipsilateral side in both gaits. All the observed changes in the activation for each phase indicated a longer duration of activation of the stroke group. It was not possible to detect reliable median frequency reduced values. **Conclusions:** ES remains the same behavior in post-stroke individuals, when compared to healthy group, however there was asymmetry between the sides. The walker did not affect the contralateral ES muscle pattern, but ipsilateral ES muscle was more activated with gait assistance.

Keywords: Stroke; Erector Spinae; Gait; Fatigue; Asymmetry.

5.2. INTRODUCTION

Stroke has been considered one of the most common causes of walking disabilities worldwide (VAN KAMMEN et al., 2017), reducing the independence of individuals affected due to their lack of ability in performing many daily tasks, and resulting in physical, psychological and economic problems (BELDA-LOIS et al., 2011). Hemiparesis, muscle spasticity and poor balance are some of the clinical features in post-stroke individuals (CAPÓ-LUGO; MULLENS; BROWN, 2012).

Trunk function is one of the main factors to be observed after stroke, however, its muscles' activities have been little studied in gait analysis. Damages in these muscle functions affect the patient's mobility, posture and balance (PEREIRA et al., 2011), increasing the rate of falls (KARTHIKBABU et al., 2012). In addition, transition from sit to stand position, typical during daily tasks, is also affected (BOUKADIDA et al., 2015).

Trunk muscles are innervated by ipsilateral and contralateral hemisphere, whereas distal muscles are innervated mainly by the contralateral one (DICKSTEIN et al., 1999; FUJIWARA et al., 2001). After a stroke, the trunk function is affected bilaterally (DICKSTEIN et al., 2004; GJELSVIK et al., 2014; KARTHIKBABU et al., 2012). Contrarily to what happens with the limbs, even if both sides are injured after stroke, trunk functions are less affected than the limbs, due to bilateral innervation (DICKSTEIN et al., 2004). Using Surface Electromyography (sEMG), decreased trunk muscle activation in both sides can be observed, whereas the muscle pattern of lower and upper limbs is impaired mainly in the contralateral side (KARTHIKBABU et al., 2012).

Due to the trunk muscle functions, they have arisen as an alternative for gait analysis and control of robotic orthoses, since their activities may be more preserved in some diseases as stroke. Additionally, the signal acquisition is more comfortable for the patients and, furthermore, there is the possibility of assessing their posture during the rehabilitation sessions. The information provided by sEMG may also anticipate propulsive phases in gait with a cycle pattern (DELISLE-RODRIGUEZ et al., 2015). However, few studies have addressed trunk muscles activities for both gait analyses and control of robotic orthoses.

On the other hand, in post-stroke individuals, the fatigue occurs earlier than in healthy individuals. Therefore, neuromuscular fatigue can be a limiting factor during the rehabilitation process. When the individual has fatigue, the training session must be interrupted in order to the patient to recover and then continue. However, depending on the duration of each session, the recovery may not be carried out, and the individual cannot complete it (BOUDARHAM et al., 2014; DUNCAN et al., 2015; XU; CHU; ROGERS, 2014).

Van Criekinge et al. (2017) have claimed that studies examining trunk muscle activity during walking in post-stroke patients are lacking, and to the extent of our knowledge, no study involving muscle analysis was performed in post-stroke patients in free and walker-assisted gaits.

Thus, the objective of this study is to compare the activity and the fatigue of trunk muscle and lower-limb muscles during free and walker-assisted gait in post-stroke individuals. We also will verify the activation symmetry and neuromuscular fatigue of erector spinae (ES) and rectus femoris (RF) in both sides of the body after stroke. We believe that the identification of these parameters can contribute for lower-limb's post-stroke rehabilitation. For instance, using the activity of trunk muscles and the neuromuscular fatigue can be used to control rehabilitation devices in physical activities during therapy sessions.

5.3. MATERIAL AND METHODS

5.3.1. Participants

Ten post-stroke (5 female and 5 male, aged 32-59 years) and 30 healthy (15 female and 15 male, aged 18-38 years) volunteers participated of this cross-sectional observational study. All volunteers signed the Free and Informed Consent Form and the study followed the ethical aspects in research with humans, being approved by the Ethical Committee of Federal University of Espirito Santo (UFES/Brazil), number CAAE: 64797816.7.0000.5542.

As inclusion criteria for the stroke group, the subject should:

- have had suffered a stroke resulting in hemiparesis;
- have suffered a stroke at least 6 months before the experiment;
- be in the level ≥ 2 according to the Functional Ambulation Category (FAC) (ANNEX B) scale (HOLDEN et al., 1984), which is normally used to assess the level of human assistance during walking;
- have enough cognitive and language skills to understand and follow the instructions about the experiment.

The exclusion criteria were individuals unable to walk independently or with some locomotor damage (lower-limbs and trunk) unrelated to stroke.

For the control group, the inclusion criteria were having no motor impairment or pain in the trunk or lower-limbs and having enough cognitive skills and language for following the experiment instructions.

5.3.2. Clinical evaluation

In the clinical trials, the anthropometrical (height, weight and BMI) data were recorded, as well as, gender and age for both groups, as comorbidities and specific features of each subject can influence the results of gait analysis.

All volunteers were recruited by an occupational therapist of our research group. In the stroke group, the volunteers aged 32-59 years. This range is important for the present study, as our experiments require the subject to walk, and adults usually recover gait faster and more efficiently than elderly (ALAWIEH; ZHAO; FENG, 2018; LUI; NGUYEN, 2018).

Before the experiments, information of subject’s medical history was requested, including: type of stroke (ischemic or hemorrhagic), time after stroke (months), side of the brain lesion, comorbidities and use of medicines. Additionally, an anamnesis form was applied, in which we confirmed the information from the medical reports, asked about dominant side, history of the stroke, alcohol consumption and smoking before and after stroke, stroke sequelae and use of assistive devices to walk.

Spasticity was evaluated using Ashworth's Modified Scale (ANNEX D), which tests resistance to passive movement and scores the muscle spasticity in patients with neurological conditions (BOHANNON; SMITH, 1987). This scale varies from 0 (no increase in tone) to 4 (affected parts rigid in flexion or extension), with intermediary grades 1, +1, 2 and 3. The Functional Ambulation Classification (FAC), described by Holden et al. (1984), was used to evaluate the amount of human assistance, rather than devices, required for ambulation. It has 6 scores, which vary from 0 (nonfunctional ambulation) to 5 (ambulator independent). Other specific features of each subject were recorded and all this information is presented in Table 6 and Table 77.

5.3.3. Experimental setup

For comparison between control and stroke groups, the trunk muscle ES on level of cervical vertebra C7, thoracic vertebra T12 and lumbar vertebra L4, and lower-limb muscles responsible mainly for knee flexion/extension — biceps femoris (BF), rectus femoris (RF), and vastus lateralis (VL) — were analyzed through sEMG. On the other hand, the accelerometer was positioned above the lateral malleolus, and the reference electrode over the medial malleolus (Figure 17).

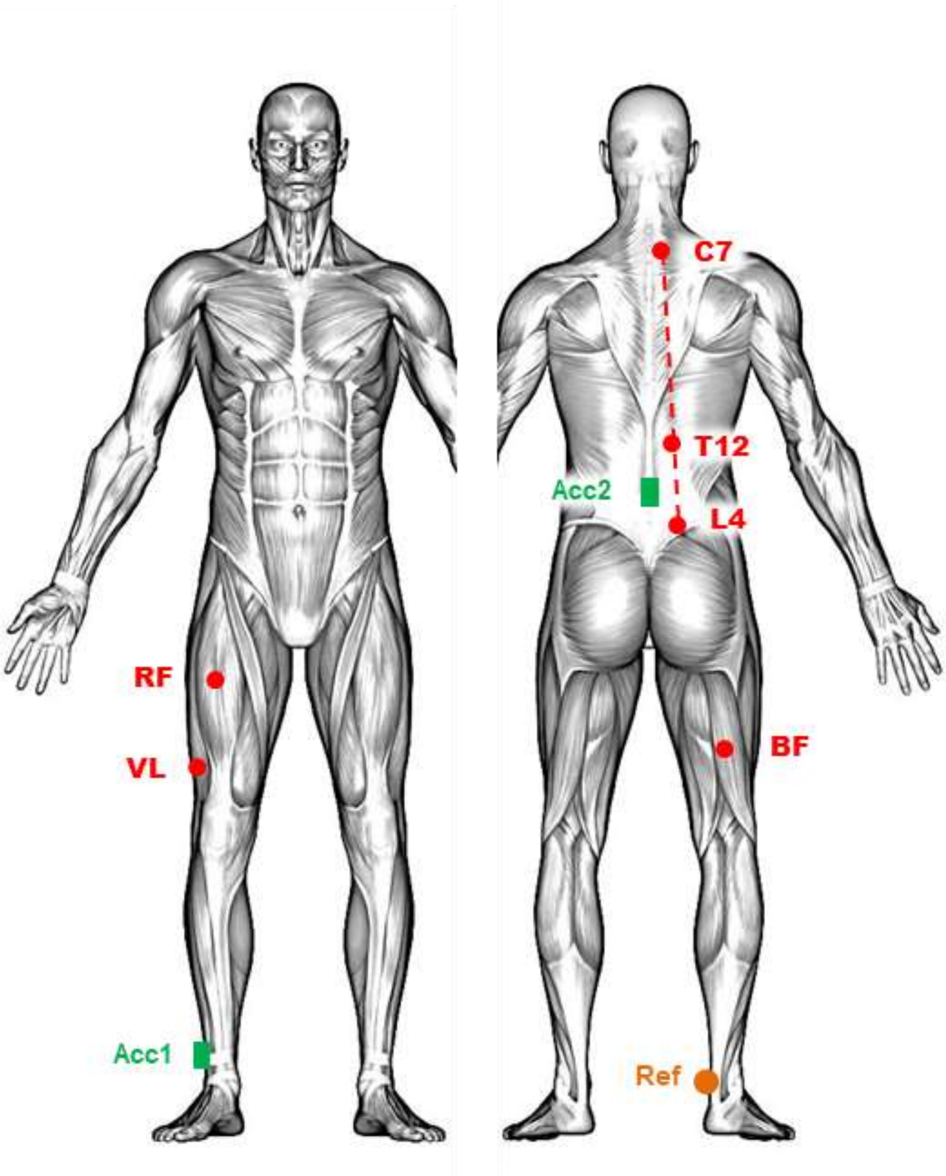

Figure 17. Positions of sEMG electrodes and accelerometer sensor. C7, T12 and L4 are the erector spinae levels analyzed; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis; Acc1: accelerometer position during the first stage of the experiment; Acc2: accelerometer position during the first stage of the experiment; Ref: reference electrode.

In this experiment, the muscles of the right side of body of the control group were analyzed. On the other hand, for the stroke group, the contralateral side to brain lesion was the one analyzed. The participants of both groups were asked to:

- Walk as he/she walks normally, but without assistive devices for a path of 10 meters, with comfortable speed and for three times;

- Walk using a modified conventional walker with wheels, adjusting its height to maintain his/her body as erect as possible for a path of 10 meters, with comfortable speed and for three times (Figure 18).

Figure 18. Volunteer, named as P2, performing the experiments of the first stage. On the left, she walks without assistance, and, on the right, she walks assisted by the modified conventional walker.

Due to channel number limitation of the sEMG acquisition equipment, the stroke group performed a second stage. Here, both ipsilateral and contralateral sides were analyzed and, this time, the sEMG signals of three muscles on each side were captured: T12, L4 and RF, in order to compare the symmetry of muscle activation in these volunteers. The accelerometer was positioned on the L2 vertebra (on the back of the subject), and the reference electrode on the medial malleolus. The volunteers of stroke group were asked to walk, as described in the first stage, performing not-assisted and assisted gaits.

5.3.4. Data collection

Muscle activity and acceleration data were recorded simultaneously using an acquisition equipment EMG 830C (EMG System do Brasil Ltda®) with 16-bit analog/digital conversion resolution, which has 6 sEMG channels and a biaxial accelerometer input. The sampling frequency used was 1000 Hz.

For allocation of electrodes on the lower-limb muscle (BF, RF and VL), we followed recommendations from Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM, 2016), and, for the ES muscle (C7, T12 and L4), the position was determined according to studies of De Sèze et al. (2008) and Swinnen et al. (2012). Before the electrode placement, the skin was cleaned (alcohol 70%) to reduce impedance. We used bipolar electrodes (Ag-AgCl, pre-gelled, 25 mm of inter-electrode distance and 10 mm of diameter). The accelerometer was positioned above the lateral malleolus and on the L2 with the y-axis pointing vertically and x-axis pointing anteriorly.

5.3.5. Data analysis

5.3.5.1. Gait phases identification

The acceleration data was analyzed using a custom algorithm developed in MatLab (2016a), which detects specific points of the signal, representing the following temporal events: initial foot contact (heel-strike) and terminal foot contact (toe-off). Thus, it is possible to divide the gait in stance and swing phases. These points were confirmed by visual analysis and, after confirmation, the signal was cut in toe-off of the right (control group) or contralateral side (stroke group), and the sEMG signals were cut in the same points. These data were normalized as a percentage of gait cycle (0 to 100%) and their means were calculated.

Some studies (BEN MANSOUR; REZZOUG; GORCE, 2015; HAN et al., 2009; LEE et al., 2010) used accelerometer on the ankle (or distal edge of the shank) in healthy subjects. The vector module of the x and y axis of the accelerometer used in this study was calculated, and two characteristic peaks were identified, in which it was possible to identify the initial foot contact (heel-strike) and terminal foot contact (toe-off). As far as our knowledge, there is only one study found in the literature (SAREMI et al., 2006), which used an accelerometer to collect kinematic parameters in stroke gait individuals, finding the same pattern of peaks of healthy gait volunteers.

For identification of heel-strike and toe-off from the L2 position, we used anterior-posterior and vertical acceleration data, respectively, as done by Zoffoli et al. (2017) and Ben Mansour et al. (2015). On the other hand, to differentiate ipsilateral and contralateral steps, we asked to participant to begin the walk using the contralateral lower limb.

5.3.5.2. Onset/offset identification

After cutting sEMG signals in gait cycles using acceleration data, they were full-wave rectified, filtered using a fourth order Butterworth band-pass filter with cut-off frequencies of 10 and 500 Hz, and normalized using the method of finding the maximum peak during the movement. This is considered the best method for dynamics contractions analyses and for neurologic patients as they are not able to produce a reliable maximum contraction (PERRY; BURNFIELD, 2010). The data was then converted to envelopes by the Root Mean Square (RMS) technique. Finally, the k-means clustering technique was used to identify the muscle onset and offset (DEN OTTER et al., 2007).

5.3.5.3. Neuromuscular Fatigue Identification

The neuromuscular fatigue was analyzed for the stroke group to verify if there were differences between ipsilateral and contralateral sides. Using only the data from the second stage, five sequential gait cycles of each trial were selected to be analyzed and, using the filtered sEMG signals, their Median Frequency (MDF) of the power spectra was determined by short-time Fast Fourier Transform (FFT) with windows of

256 ms, such as done in (COOREVITS et al., 2008). As fatigue accumulates, the activation energy becomes lower. Therefore, it is expected that the MDF of sEMG reduces after gait (KIM et al., 2013).

The regression coefficient of the MDF slope towards lower frequencies can be used as a non-invasive fatigue index for the investigated muscle (KONRAD, 2005). To calculate the decrease of the MDF, a linear regression of the signal was computed and plotted in the same graphic as a continuous transition. The first point of the line was considered as reference (100%), and the end point was calculated, in percentage, with respect to the initial point.

5.3.6. Statistics

All the data from control and stroke group were processed using descriptive statistics, calculating mean and standard deviation (SD) for each people and group. The coefficient of variation (CV), which is calculated as (SD/ mean) x 100, was used to verify how much the values varied within a sample. Samples with CV values lower than 15% were considered to have low dispersion; from 15 to 30%, moderate dispersion; and higher than 30%, high dispersion.

Using the Shapiro-Wilk normality test, we identified that the samples were not normally distributed. Therefore, the Mann-Whitney test (nonparametric test for two independent samples) was applied to verify statically significant differences between control and stroke groups. The Wilcoxon signed-rank test (nonparametric test for paired samples) was applied to compare the free and assisted gaits in the stroke group, ipsilateral and contralateral muscle activation, and neuromuscular fatigue in the stroke group. When p-value was < 0.05, it was considered that there was statistical difference between the two samples.

The number of volunteers (stroke group = 10) generated a sample size for this study that has a moderate effect size of $|\rho| = 0.69$ (AGUIAR et al., 2018), with statistical power of 75% and alpha = 0.05.

Table 6. Post-stroke individuals' information.

Subject	Gender	Age (years)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Type of stroke	Time after stroke (month)	Side of the brain lesion	Paretic side	Dominant side	FAC	Ashworth's scale
P1	F	44	164	73	27	H	61	L	R	R	4	1
P2	F	32	165	65	26	H	22	L	R	R	2	1+
P3	F	48	165	68	25	I	8	L	R	R	3	2
P4	M	59	171	82	28	I	8	R	L	R	4	1+
P5	M	48	175	63	21	H	6	R	L	R	3	2
P6	F	33	170	65	26	I	7	L	R	R	4	1+
P7	M	54	174	71	24	I	18	R	L	R	5	2
P8	M	58	168	68	24	I	12	R	L	R	3	2
P9	M	43	169	64	22	I	6	R	L	R	5	2
P10	F	55	155	66	28	I	6	R	L	R	5	1
Mean ± SD	5M/5F	47 ± 9	168 ± 6	69 ± 6	25 ± 2	7I/3H	15 ± 17	6R/4L	6L/4R	10R	4 ± 1	-

F: female; M: male; H: hemorrhagic; I: ischemic; L: left; R: right; FAC: Functional Ambulation Category; BMI: Body Mass Index; SD: Standard Deviation.

Table 7. Description detailed from each post-stroke individual.

P1	She suffered a hemorrhagic stroke due to injury during a convulsive crisis. Although she suffered stroke more than 5 years ago, she presented spasticity, characteristic hemiparetic gait and a mild cognitive deficit.
P2	She suffered a hemorrhagic stroke due to a motorcycle accident, had mild speech difficulties and reported frequent tiredness. She showed more difficult to walk than the others subjects.
P3	She had hypertension and diabetes.
P4	He had hypertension and dyslipidemia.
P5	He suffered a hemorrhagic stroke caused by an unreported injury and has used antiepileptic medication.
P6	She had expressive aphasia.
P7	He had hypertension and reported consumption of alcohol and smoking before stroke.
P8	He did not present specific features.
P9	He has a well-developed and preserved musculature, even in the paretic lower limb. He was the only one of the 10 participants who did not use an assistive device to walk, even presenting hemiparetic gait.
P10	She had a mild stroke (she did not need to be hospitalized), which resulted in a little spasticity in the limbs. Even using a cane, her gait pattern was more symmetrical (visual analysis) than the other participants.

All participants, except for P9, habitually used walking canes. All individuals were being treated with medicines in the time of the experiments.

5.4. RESULTS AND DISCUSSION

This study was composed of a group of 10 post-stroke individuals, and their characteristics and information are presented in Table 6 and Table 7. Additionally, a control group formed by healthy adults (15 females, 15 males, 27 ± 5 years, 169 ± 10 cm height, 67 ± 15 kg weight and Body Mass Index: 23 ± 4 kg/m2) participated of the experiments, as reference data.

5.4.1. Kinematic Parameters

5.4.1.1. Speed

For both groups was asked the subjects to walk in a self-selected speed, both in the free and assisted gaits, to allow a more natural gait possible. Table 88 shows the average speed of each gait and group. Comparing the speed of the same group in two different walks, we identified that only the control group had significant differences, being the speed of free gait (0.99 ± 0.11 m/s) higher than the assisted gait (0.88 ± 0.12 m/s) in this group. The use of the walker did not affect the speed of stroke gait.

Table 8. Speed (mean and standard deviation) of the volunteers during the experiments.

		Control Group	Stroke group	p-value (inter-group)
Speed (m/s)	Free gait	0.99 ± 0.11	0.51 ± 0.26	< **0.01**
	Assisted gait	0.88 ± 0.12	0.51 ± 0.20	< **0.01**
	p-value (intra-group)	< **0.01**	0.99	

Data in which statistically significant differences (p-value < 0.05) were found are highlighted in bold.
Wilcoxon and Mann-Whitney tests were used for intra-group and inter-group comparison, respectively.

When compared the speed between control and stroke group, both walking tasks were lower in the stroke group. Regarding the CV, there was a high dispersion for the stroke group (free gait CV = 50.1% and assisted gait CV = 40.0%), whereas the control group presented low dispersions (free gait CV = 11.2% and assisted gait CV

= 13.4%). In addition, the speed in the stroke group varied from 0.19 to 1.05 m/s during free gait, and from 0.22 to 0.80 m/s during the assisted gait. In both cases, the slowest speed was performed by participant 2 (P2) and the fastest by P10.

The control group walked slower in the free gait than data from the literature (~ 1.3 m/s) (PERRY; BURNFIELD, 2010; VERMA et al., 2012). On the other hand, Suica et al. (2016) analyzed 19 healthy subjects (22 to 70 years) and found the speed of 1.41 ± 0.15 m/s and 1.39 ± 0.15 m/s during walking without and with the use of a rollator, respectively, and there was no statistically significant difference between them.

Stroke gait speed may vary widely, according to some authors (BALABAN; TOK, 2014; VERMA et al., 2012), from 0.10 to 1.00 m/s, depending on the different levels of motor damages, age, comorbidities and time after stroke. The speed values obtained in our study for the stroke group are similar to Barroso et al. (2017), 0.52 ± 0.18 m/s, whose study included 9 post-stroke individuals classified as 4 or 5 in the FAC, and mean age of 53 years (being two elderly people).

Recovering walking ability to perform daily activities is one of the main objectives in a stroke rehabilitation process (AGUIAR et al., 2018). Low velocities may indicate disabilities and difficult outdoor tasks; however, post-stroke individuals may increase it by developing compensatory mechanisms, which may be harmful (BARROSO et al., 2017; BEYAERT; VASA; FRYKBERG, 2015). These mechanisms usually generate an asymmetric gait pattern (kinematic, kinect and muscle activation) and overload the ipsilateral side.

In our control group, there was only one significant difference: a reduction in the speed during the walker-assisted gait, which was expected, according to the literature (MARTINS et al., 2012). Nevertheless, this reduction did not affect the phases duration because the toe-off in the assisted gait remained with similar value to the free gait. Moreover, no muscle had significant alterations in its onset/offset.

5.4.1.2. Accelerometer and cycle phases

In the first stage of the experiments, an accelerometer was used above the malleolus lateralis to identify gait phases of the right lower limb in the control group and contralateral lower limb in the stroke group. The heel strike was identified and used as reference of 0 and 100% of the gait cycle, and the toe-off divides it in stance and swing phase, e.g., the duration of the stance phase occurred from 0 to toe-off, and the swing phase is from toe-off to 100%. The toe-off percentages are presented in Table 99, where it is possible see that the walker did not modify the stance duration in both groups, and the stroke group had a stance phase smaller in both gaits when compared to the control group. The duration of stance phase of the healthy gait matches the literature, which states it is about 62% of the gait cycle (PERRY; BURNFIELD, 2010).

In the second stage, the goal was to verify the assymetry between both contralateral and ipsilateral sides of the post-stroke subjects. In this case, the accelerometer was positioned on the back, where heel strike and toe-off were identificated in both lower limbs.

The mean toe-offs were 56.1 ± 3.0% (Free Gait - Contralateral side), 55.7 ± 2.7% (Assisted Gait - Contralateral side), 65.8 ± 6.2% (Free Gait - Ipsilateral side), and 64.9 ± 6.1% (Assisted Gait - Ipsilateral side). The p-values calculated in the comparison between contralateral and ipsilateral sides were $p < 0.01$ during free gait and $p < 0.01$ during assisted gait. In comparison between free and assisted gaits, it was $p = 0.64$ for contralateral side and $p = 0.51$ for ipsilateral side. Therefore, the ipsilateral limb had a longer stance phase than the contralateral in both gaits, which was expected (ALLEN; KAUTZ; NEPTUNE, 2011), and the walker did not modify the duration of the phases in this group.

Studies conducted by Lamontagne et al. (2000) and Den Otter et al., (2007) found the following durations of the stance phase in stroke gait, respectively, contralateral of 67% and ipsilateral of 74% (30 subjects at 0.48 m/s), and contralateral of 69% and ipsilateral of 68% (24 subjects at 0.35 m/s). The first one identified that ipsilateral limb had longer stance phase, such as found in our results, however, the values were very different. Meanwhile, the second study obtained similar durations, both longer than the healthy gait.

Temporal asymmetry is a common characteristic of the hemiparetic gait, usually showing a reduced duration of stance phase of the contralateral side. Due to the body stability needed for walking, the ipsilateral limb is overloaded in free gait, as it has a higher stance phase and remains more time supporting the body weight (ALLEN; KAUTZ; NEPTUNE, 2011; DOBROVOLNY et al., 2003). Furthermore, a high asymmetry is associated with a slower self-select speed (LEWEK et al., 2014). For the reason, post-stroke individuals usually walk with assistance devices, for example walker and canes, to provide an improvement in symmetry, relieve load on the ipsilateral limb, and provide stability and balance (VERMA et al., 2012).

5.4.2. Muscle activation

5.4.2.1. Control and stroke groups

In Table 99, the means of muscle onset/offset of both groups in free and assisted gait are presented. ES muscle was analyzed in three levels, and we observed that the upper level (C7) was previously activated, followed by T12 and L4, respectively. The activation pattern of the ES muscle followed a similar pattern to the RF muscles. Therefore, ES muscle had two activation periods: the first one begins at pre-swing (stance phase) and ends at initial swing (swing phase). The second period begins at terminal swing (swing phase) and ends at the mid stance (stance phase). The ES muscle was activated previously to the RF, except for the onset in the swing phase, where RF onset occurred near T12 and L4 onsets.

Some studies (ANDERS et al., 2007; CECCATO et al., 2009; WHITE; MCNAIR, 2002; ZOFFOLI et al., 2017) have analyzed the ES muscle during gait in healthy subjects and all of them found an ES activity preceding the lower limb muscles activities, in addition to a sequential recruitment of ES levels, where the upper level begins before than the others levels and so on. An important work (CECCATO et al.,

2009) analyzed five ES levels (C7, T3, T7, T12 and L3) in healthy men, identifying two activation peaks, one in the first double support (~ 0-10% of the gait cycle) and another, more prominent, in the second double support (~ 50-60%). Similar results were obtained by White and McNair (2002).

The knee flexors/extensors have been quite studied during human gait. Many authors (CRIEKINGE et al., 2018; PERRY; BURNFIELD, 2010; WARD et al., 2018; WHITTLE, 2007) presented the most common pattern for these muscles. The pattern obtained in our study during healthy gait was similar to the literature. All three muscles (BF, RF and VL) have an activation beginning in the final of swing phase, which extends to ~25% of the cycle, and RF presents one more activation, which concentrates around the toe-off (onset in the final of stance phase and offset in the begin of swing phase).

Comparing muscle activation of control and stroke groups, there was no significant difference only in the T12 onset in the swing phase and in the T12, L4 and RF offsets in the stance phase, in the free gait, and in C7 and BF onsets, in the swing phase, C7 and RF offsets, in the stance phase, and T12 offset in the swing phase. All other onset/offset presented significant differences.

In the free gait of the stroke group, the ES muscle activation near the toe-off was longer in the three levels, when compared with the control group. This same feature was observed in the comparison between groups during the assisted gait. A longer double support before the paretic limb swing may be the cause of these alterations.

During free gait, BF (stroke group) activation began later and remained for the most part of stance phase, ending in 54.8 ± 15%, differently of the control group. In the assisted gait, also the BF onset began later, however, the offset occurred before, at 35.8 ± 4.8% (significant difference in the BF offset using walker), and even so it was later than the control group in the assisted gait. In order to avoid knee hyperextension caused by quadriceps spasticity, BF showed a higher activation time, and, therefore, there was more coactivation between quadriceps and hamstrings muscle groups (CORRÊA et al., 2005).

Suica et al. (2016) analyzed the rollator-assisted gait in healthy subjects and identified a reduced muscle activity of RF and semitendinosus (hamstring muscle, such as BF) caused by the weight bearing imposed on the walker. Our results from the control group did not present significant differences; however, we observed a reduction in the BF activation in the stroke group.

In both gaits, VL onset of the stroke group occurred earlier and this muscle had a longer duration of activity in the stance phase, compared with control group.

5.4.2.2. Contralateral and ipsilateral sides of the stroke group

One of the main characteristics of post-stroke gait is the asymmetry caused by the hemiparesis, which affects not only the lower limb, but also trunk kinematics and arms swing (JOHANSSON et al., 2014). These unilateral motor dysfunctions contribute to reduce the balance in post-stroke subjects (KARTHIKBABU et al., 2018). As a way of circumventing the spasticity and muscle weakness caused by stroke, the human organism developed compensatory mechanisms for performing the gait. These mechanisms usually generate increased energy expenditure for an individual during walking, and the ipsilateral limb plays an extra function, as its stance phase is longer than contralateral limb, supporting a higher load, and the contralateral limb has more muscle coactivation to counterbalance the effect of spasticity (THIJSSEN et al., 2007).

Table 9. Comparison of muscle activation among control (right side) and stroke (contralateral side) groups, and free and assisted gait in stroke group.

		Free Gait			Assisted Gait			Free x Assisted Gait (Stroke Group)
		Control Group (mean ± SD)	Stroke Group (mean ± SD)	p-value (Mann-Whitney's test)	Control Group (mean ± SD)	Stroke Group (mean ± SD)	p-value (Mann-Whitney's test)	p-value (Wilcoxon's test)
Toe-off (%)		61.6 ± 2.8	56.1 ± 4.6	< **0.01**	61.4 ± 2.9	54.9 ± 2.8	< **0.01**	0.76
ONSET (% gait cycle)	C7	36.0 ± 3.1	28.3 ± 1.5	**0.01**	35.7 ± 2.7	25.0 ± 5.3	**0.01**	0.08
		81.9 ± 3.1	88.2 ± 0.7	**0.03**	81.7 ± 2.2	83.0 ± 3.1	0.30	0.15
	T12	48.4 ± 2.8	36.6 ± 7.9	< **0.01**	48.9 ± 2.9	31.6 ± 4.0	< **0.01**	0.16
		93.4 ± 2.9	89.5 ± 8.1	0.08	90.1 ± 3.3	85.1 ± 3.5	**0.01**	0.29
	L4	51.6 ± 2.5	38.1 ± 8.0	< **0.01**	50.4 ± 2.6	37.2 ± 5.9	< **0.01**	0.68
		96.3 ± 2.9	89.8 ± 7.1	**0.01**	95.3 ± 3.2	87.6 ± 4.2	< **0.01**	0.50
	BF	81.2 ± 2.8	87.8 ± 6.6	**0.01**	84.0 ± 2.5	87.1 ± 7.8	0.21	0.97
	RF	56.6 ± 2.9	52.2 ± 4.5	**0.02**	55.3 ± 3.2	42.9 ± 9.8	< **0.01**	**0.04**
		92.8 ± 2.6	87.2 ± 5.9	**0.01**	93.9 ± 2.7	87.9 ± 4.6	< **0.01**	0.83
	VL	93.1 ± 2.9	86.8 ± 5.9	< **0.01**	92.5 ± 3.1	88.3 ± 5.7	**0.03**	0.52
OFFSET (% gait cycle)	C7	62.4 ± 3.1	68.8 ± 6.5	**0.02**	59.6 ± 3.3	63.3 ± 7.4	**0.01**	0.66
		9.8 ± 2.8	3.0 ± 0.7	**0.03**	5.4 ± 2.9	5.0 ± 2.3	0.79	0.15
	T12	65.6 ± 2.8	71.5 ± 5.8	**0.02**	67.2 ± 3.0	66.9 ± 7.9	0.81	**0.04**
		11.9 ± 3.2	10.5 ± 9.3	0.29	12.8 ± 2.4	8.3 ± 3.3	< **0.01**	0.79
	L4	66.1 ± 2.7	75.3 ± 12.2	< **0.01**	65.7 ± 3.1	69.5 ± 4.7	**0.01**	0.08
		17.9 ± 3.1	15.9 ± 10.1	0.78	18.6 ± 2.7	14.5 ± 4.7	**0.03**	0.82
	BF	23.1 ± 2.8	54.8 ± 15.0	< **0.01**	23.8 ± 3.2	35.8 ± 4.8	< **0.01**	< **0.01**
	RF	72.3 ± 2.8	62.3 ± 5.4	< **0.01**	73.8 ± 3.1	65.4 ± 5.9	< **0.01**	0.12
		25.0 ± 2.9	24.1 ± 8.3	0.65	25.6 ± 3.2	26.7 ± 10.5	0.23	0.49
	VL	25.9 ± 2.4	44.6 ± 12.0	< **0.01**	24.9 ± 2.9	41.7 ± 7.8	< **0.01**	0.50

Data in which significant statistic differences (p-value < 0.05) were found are highlighted in bold.
C7, T12 and L4 are the erector spinae levels analyzed; BF: biceps femoris; RF: rectus femoris; VL: vastus lateralis; SD: Standard deviation.

Initially, the stroke may interfere in the muscle functions if the motor cortex (control the skeletal muscles activities) has been affected. Throughout the time, after a stroke, skeletal muscles have structural changes, due to atrophy related to disuse, and increased intramuscular fat and fibrous tissue, being these changes observed in both sides of the body (BERENPAS et al., 2017; RYAN et al., 2011; SCHERBAKOV; SANDEK; DOEHNER, 2015).

The changes in the trunk muscles functions after a stroke are not so easy to be detected as the limb muscle functions, since both brain cortex hemispheres innervate both sides of these muscles, which is different for the lower limbs that are innervated mainly by the contralateral brain cortex (QUINTINO et al., 2018). Van Criekinge et al. (2017) claim both sides of the trunk after stroke may present reduced muscle activity levels, delayed onset times, and diminished synchronization of the trunk muscles. Electromyography techniques to quantify and compare activation timings are important in gait analysis, mainly in post-stroke individuals, as in cyclic activities such as gait, the muscles need to produce activations and also activate them at the accurate time (ANDROWIS et al., 2018).

The second stage of this study aimed analyzing the symmetry between contralateral and ipsilateral sides. The average muscle pattern was calculated for the stroke group and is shown in Figure 19.

Initially, we compared the influence of the walker in the muscle pattern of contralateral side and observed there was no statically significant difference. Unlikely, during the assisted gait, the duration of T12 activation of the ipsilateral side was longer, because the offset was later in stance and swing phases. In addition, the RF offset was earlier in the assisted gait, indicating a smaller time of activation in the second double support.

Regarding the comparison of contralateral and ipsilateral sides, more alterations were verified. Such as aforementioned, the stance phase of ipsilateral limb was longer than the contralateral limb. Also, during the free gait, the contralateral T12 activation in the second double support (near the toe-off) was longer, whereas the other contralateral T12 activation did not present changes. Contralateral L4 onset in the stance phase occurred earlier and the duration of this activation was longer, compared with ipsilateral side, and, there was no significant difference in the other contralateral L4 activation. Both ES muscle levels presented the same alteration between contralateral and ipsilateral sides. Finally, RF offsets were later in the ipsilateral side than the contralateral side.

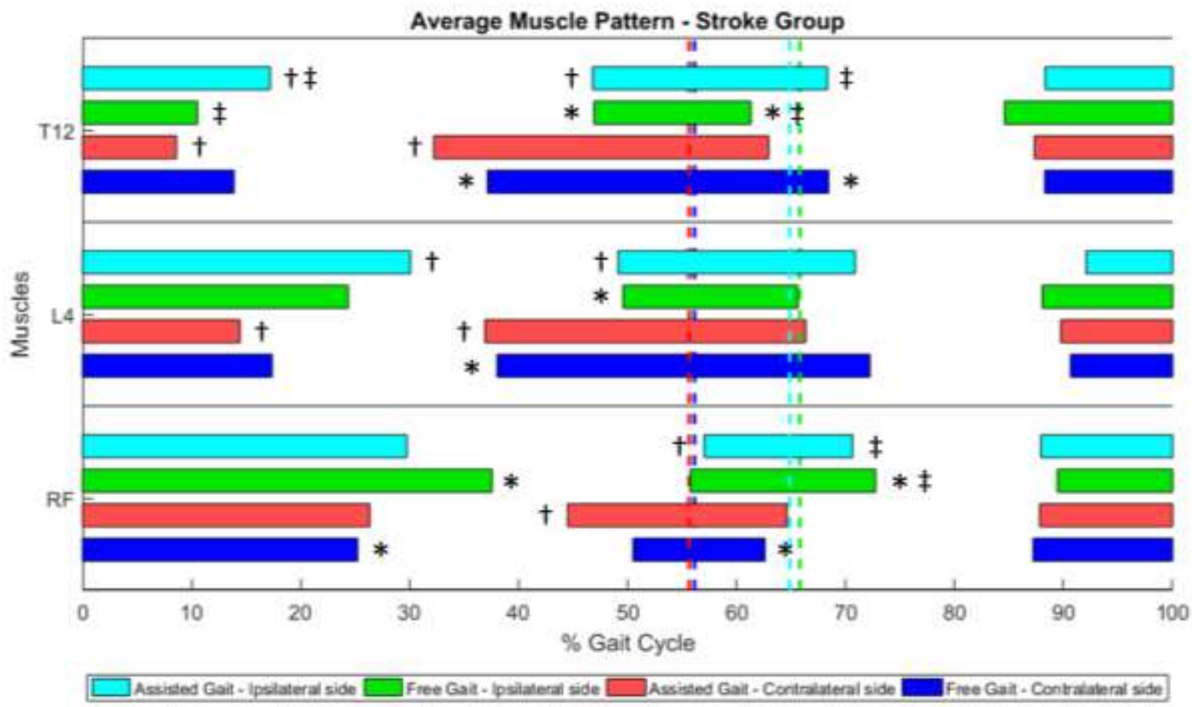

Figure 19. Average muscle pattern obtained from the stroke group in the second stage of the experiments, analyzing both sides of the body. The symbols *, † and ‡ indicate statistically significant differences in the muscle activation, where * and † were used for comparison between contralateral and ipsilateral during free and assisted gaits, respectively, and ‡ was used for comparison between free and assisted gaits in the ipsilateral side. There was no statistically significant difference between free and assisted gaits in the contralateral side. The dashed vertical line represents the toe-off of the group mean. T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

When we compared both sides in the assisted gait, it was possible to identify that the contralateral side had an earlier T12, L4 and RF onsets in the stance phase, and an earlier T12 and L4 offsets in the stance phase. Thus, using a walker, the duration of the contralateral ES muscle activation was longer in the second double support, similar to the free gait, and shorter in the first double support (beginning of the stance phase).

5.4.2.3. Post-stroke individual analysis

Usually, post-stroke individuals are heterogeneous among them due to type and local of the brain lesion, age, time after stroke, comorbidities, etc. However, there are some common general characteristics in the muscle activation during gait, as earlier onset and longer duration of activation than healthy gait (BALABAN; TOK, 2014). The means of the stroke group presented higher values of SD than control group, and the post-stroke individual characteristics shown in Table 6 and Table 7 indicate some heterogeneity in the stroke group. For this, Figure 20 shows the muscle pattern obtained for each participant and the group average to observe as the values varied. The highest variations were the patterns of P2 (the most different pattern, with a longer stance phase and, therefore, displaced periods of activation) and P3 (later ES muscle activation and earlier RF activation). Only P2 was classified as 2 in FAC, presenting the most impaired gait and was the only one related to tiredness.

5.4.2.4. Activation in gait phases

Additionally to identify activation timings during a gait cycle, other important measure in muscle activity analysis is the proportion of time the activation lasts in each gait phase. Considering each phase as 100% and calculating how much time the muscle remained active in it, we can verify if a gait modality requires more effort than other gait, or if a side of body is more overloaded during a task (DEN OTTER et al., 2007; LAMONTAGNE; RICHARDS; MALOUIN, 2000). Figure 21 presents the percentage of activation of each muscle during the stance and swing phases of the stroke group. Calculating CV for these data, we found mainly moderate dispersion in the stance phase values and high dispersion in the swing phase.

In the contralateral side, the use of walker did not alter the time of ES muscle activation during stance phase; however the time of RF activation was increased in the assisted gait. The ipsilateral ES muscle was more activated and RF was less activated with gait assistance in the stance phase, but in the swing phase there was no change.

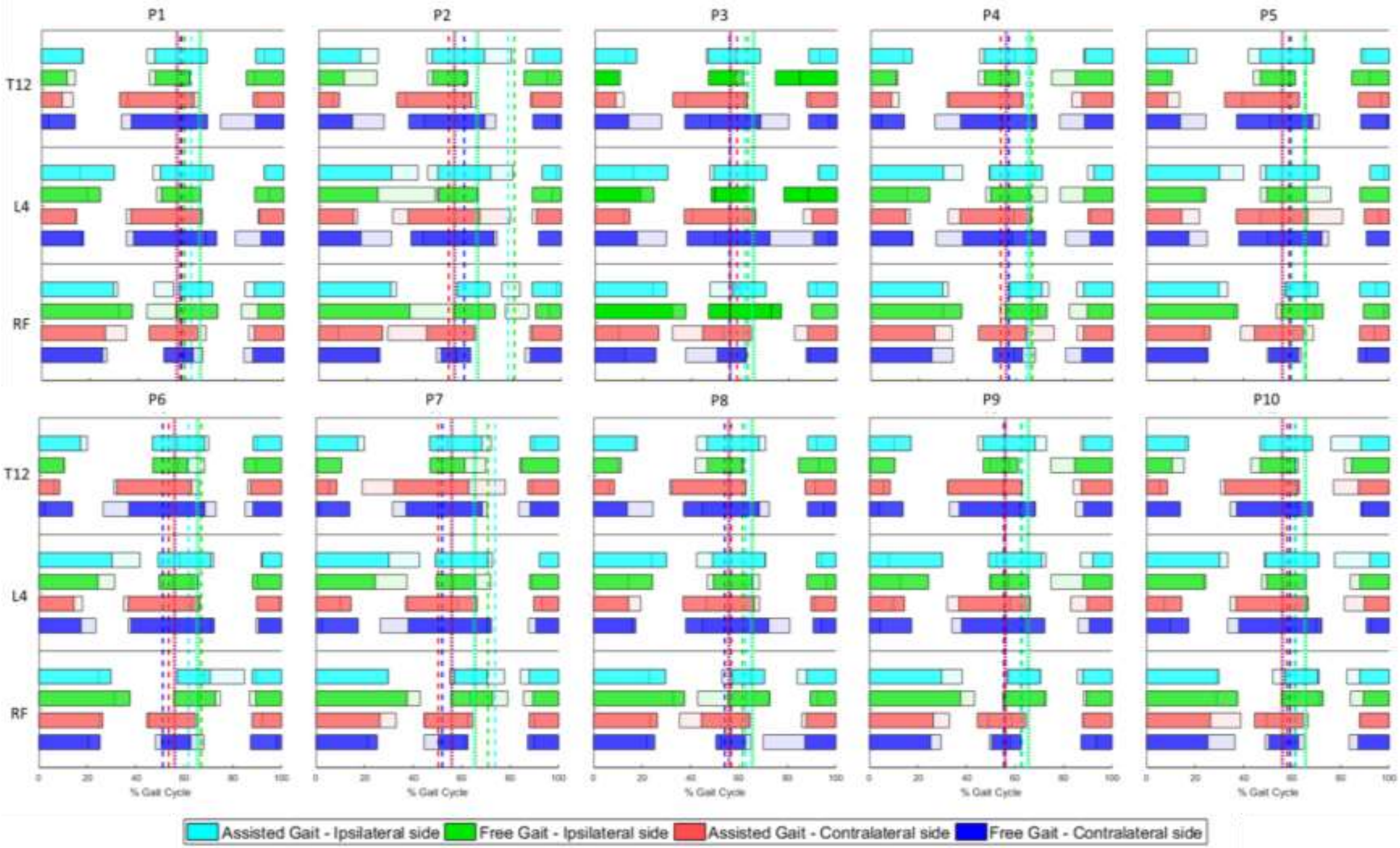

Figure 20. Average muscle patterns are presented individually for each participant, including the stroke group average to visualize as the variation occurred within the group. The dotted vertical line represents the toe-off of the group mean, and the dashed vertical line represents the toe-off of each participant. T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

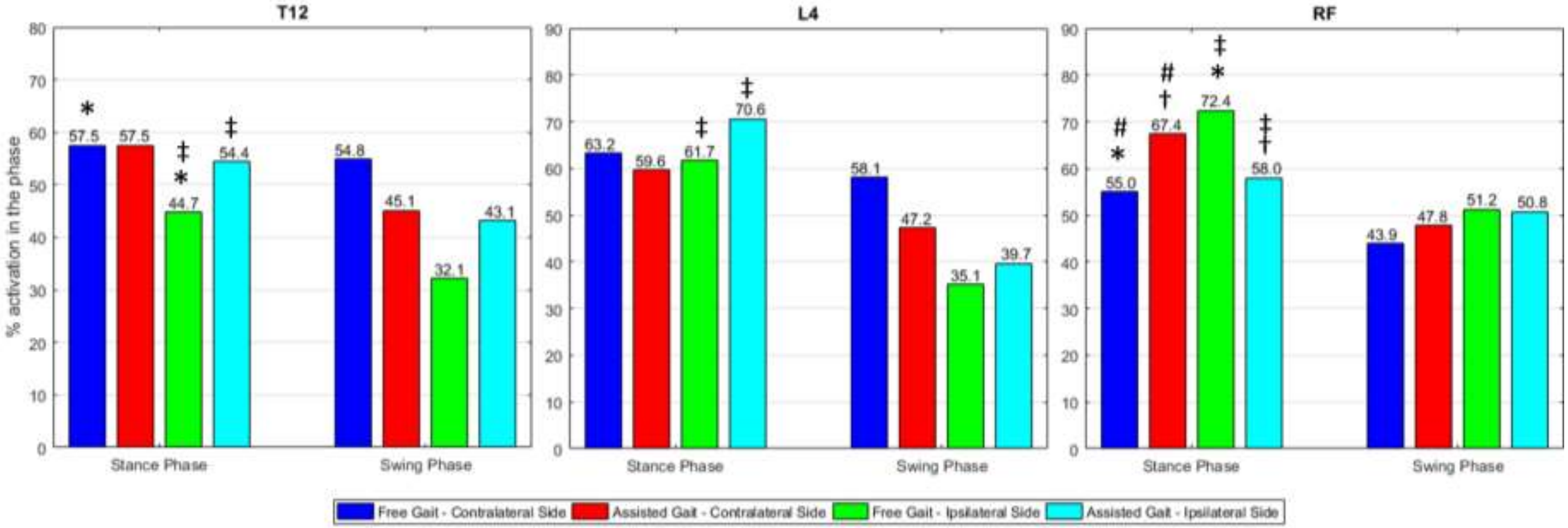

Figure 21. Percentage of activation of each muscle during the stance and swing phases. Each phase was considerate as 100% to verify how long the muscle kept activated during that phase. The symbols *, †, ‡ and # indicate statistically significant differences in the percentage in the muscle activation, where * and † were used for comparison between contralateral and ipsilateral during free and assisted gaits, respectively, and ‡ and # were used for comparison between free and assisted gaits in the ipsilateral and contralateral sides, respectively. Here, the Wilcoxon test was used. T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

During free gait, the ipsilateral T12 activation was shorter and RF activation was longer than contralateral side in the stance phase. Finally, in assisted gait, the ipsilateral RF activation was shorter than contralateral.

No muscle showed significant difference in the swing phase in all comparisons, besides the values varied more than stance phase. This was caused by the high dispersion of these data and makes the results of the swing phase less reliable.

Table 10 presents the percentage of activation in each phase of the control group and compares the values with the stroke group data. Comparing these data, only the ipsilateral T12 and contralateral RF activations did not present statistically significant difference in the stance phase of the free gait. Regarding the swing phase, the RF activations did not presented significant changes in both free and assisted gaits. All the observed changes indicated a longer duration of activation of the stroke group, except for ipsilateral T12 activation in the swing phase during free gait. Den otter et al. (2007), studying the lower limb muscles of post-stroke subjects, verified a prolonged RF activity in both ipsilateral and contralateral sides when compared with healthy subjects, similar to our findings. Both contralateral and ipsilateral muscle activation presented many significant differences in relation to healthy gait.

Table 10. Percentage of activation during the gait phases of the control group, considering each phase as 100%.

	FREE GAIT		**ASSISTED GAIT**	
	Stance Phase (mean ± SD)	**Swing Phase** (mean ± SD)	**Stance Phase** (mean ± SD)	**Swing Phase** (mean ± SD)
T12	41.2 ± 2.8 *	40.7 ± 1.5 *†	40.8 ± 2.9 *†	27.8 ± 3.1 *†
L4	48.2 ± 1.2 *†	23.4 ± 3.0 *†	45.2 ± 2.8 *†	21.4 ± 3.0 *†
RF	51.6 ± 2.9 †	47.7 ± 1.4	48.7 ± 3.0 *†	46.6 ± 2.8

Data in which significant statistic differences (p-value < 0.05 – Wilcoxon test) were found are marked with (*) for contralateral side and (†) for ipsilateral side.
T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris

In the stroke group, the symmetry between sides was calculated dividing the percentages of activation in the phase (contralateral/ipsilateral), and the closer the value is to 1.0, the more symmetrical the sides. The symmetry of the stance phase was:

- T12: 1.29 in the free gait and 1.06 in the assisted gait;
- L4: 1.02 in the free gait and 0.84 in the assisted gait;
- RF: 0.76 in the free gait and 1.16 in the assisted gait;

and of the swing phase was:

- T12: 1.71 in the free gait and 1.05 in the assisted gait;
- L4: 1.66 in the free gait and 1.19 in the assisted gait;
- RF: 0.86 in the free gait and 0.94 in the assisted gait.

Thus, the walker aided to reduce the asymmetry at 5 out of 6 measures, where only the L4 activation in the stance phase was more symmetrical during free gait.

5.4.3. Neuromuscular Fatigue

Fatigue is often in post-stroke subjects, affecting almost 50% of them, which makes difficult some daily activities (DUNCAN; WU; MEAD, 2012). Regarding the neuromuscular fatigue, the most present in these individuals is the central fatigue, which is an inability to achieve the fully activation of a muscle or maintain a specified force during an activity (BOUDARHAM et al., 2014; OLSON, 2010). From the sEMG data, the neuromuscular fatigue is characterized by a reduction of the frequency content of this power density spectrum (OLSON, 2010). However, in post-stroke subjects, the contralateral side may develop a higher level of central fatigue than the ipsilateral side and healthy subjects after a task (TOFFOLA et al., 2001).

Neuromuscular fatigue was analyzed during the second stage of the experiment, aiming to compare the MDF of T12, L4 and RF of the contralateral and ipsilateral sides. The path performed was a non-fatigable task, but our goal was precisely to verify if it was possible to detect decreases in the slopes of MDF during light exercises in post-stroke subjects.

Figure 22 shows the slopes for each one of three paths walked, for all muscles and gait modalities separately. Most of the slopes obtained did not presented significant differences among the paths. There was a decrease of 12.0% ($p = 0.04$) in the ipsilateral T12 muscle during free gait; 21.4% ($p = 0.04$) in the contralateral T12 during assisted gait; 9.8% ($p = 0.004$) and 6.2% ($p = 0.03$) in the contralateral and ipsilateral L4 muscles, respectively, during free gait. In addition to these decreases, the data were not conclusive and it was not possible to detect reduced MDF.

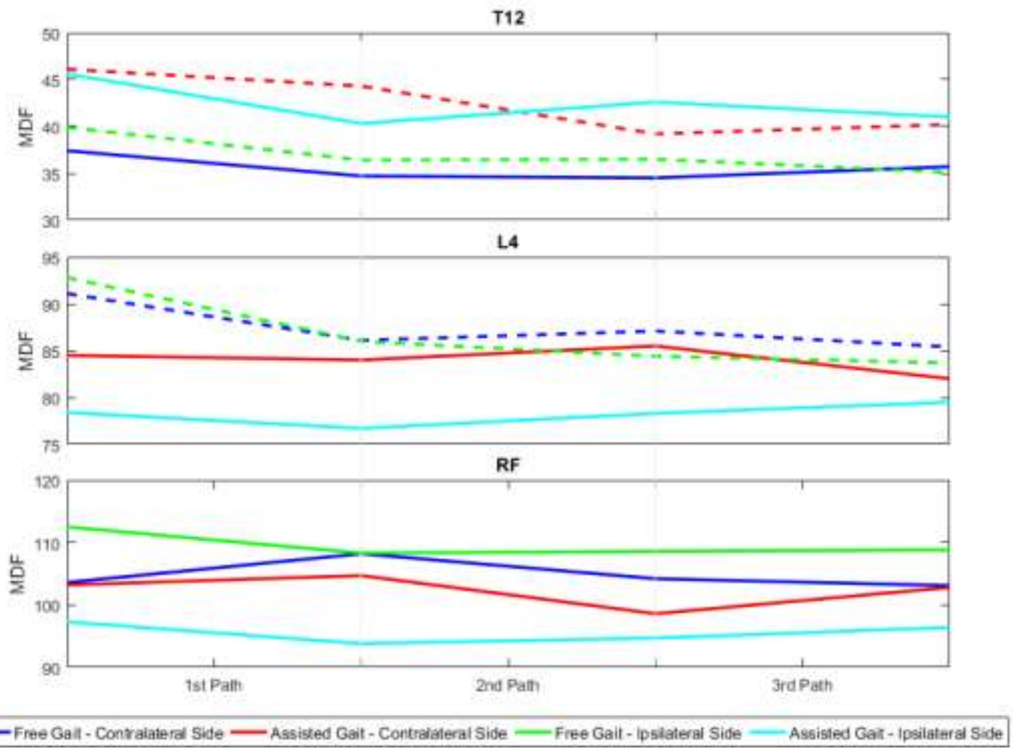

Figure 22. The dashed lines indicate there are statistically significant differences in the median frequency, which means that, in all these cases, there was decrease in MDF. T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris; MDF: Median Frequency.

5.5. CONCLUSIONS

Damages in the trunk muscle functions are a common post-stroke sequela, which may affect the individual mobility. Although the trunk function is altered bilaterally, it is more preserved than lower-limb muscle functions. Due to the importance in stability and posture during walking, trunk muscles were studied in this work in stroke and healthy subjects during two gait modalities, without and with modified conventional walker assistance.

The self-selected speed was slower among the post-stroke subjects than healthy subjects in both free and assisted gait. However, the walker did not alter the speed in the stroke group, and the speed group was relatively homogeneous.

Regarding the duration of the stance phase, both contralateral and ipsilateral limbs of stroke group were different compared with the control group, in which the stance phase was shorter in the contralateral and longer in the ipsilateral limb. Comparing symmetry between sides in post-stroke subjects, the ipsilateral limb had a longer stance phase than contralateral in both gaits and the walker did not modify the phase duration.

For both groups, ES muscle (C7, T12 and L4 levels) presented a sequential activation, beginning on the upper level, with two activation periods, near to both double supports of the gait cycle. Most of the muscle onset/onset was significantly different in the comparison between groups. ES muscle activation near the toe-off was longer, when compared to the control group, probably due to a longer double support before the contralateral limb swing.

The post-stroke subjects presented a longer BF and VL muscles activation in the stance phase than the control group in both gaits, but the use of walker reduced significantly the BF activation. The ipsilateral T12 activation was longer, and RF was shorter near the toe-off in the assisted gait. The contralateral ES muscle had longer activation near the toe-off than the ipsilateral side in free and assisted gait.

Comparing the activation in each phase of both groups, only the ipsilateral T12 and contralateral RF activations did not present statistically significant difference in the stance phase of the free gait, and all the observed changes indicated a longer duration of activation of the stroke group, except for ipsilateral T12 activation in the swing phase during free gait.

The literature suggests the contralateral side may develop a higher level of central fatigue than the ipsilateral side and healthy subjects after a task, however, in this study was not possible to detect reduced MDF, possibly due to the fact the gait task was very light.

Trunk muscle weakness after a stroke impairs balance and limits independence in walking. In this work, we assessed the timing activation, proportion of activation during stance and swing phases, and neuromuscular fatigue, during free and assisted gaits of post-stroke individuals. In addition, we verified the symmetry between contralateral and ipsilateral sides and compared muscle activity with healthy individuals.

As future work, it is important to analyze the trunk function during the rehabilitation process and verify the possibility of employment of ES signals for an application in the control of robotic devices. It is worth to emphasize that these muscles may be less affected in post-stroke individuals, have an earlier activation than lower limb muscles, and be used to assess the posture during the gait.

6. CASE STUDIES

In the Assistive Technology Group (NTA) at the Federal University of Espirito Santo (UFES), a robotic rehabilitation system composed of a robotic walker and a knee active exoskeleton is being developed (Figure 23).

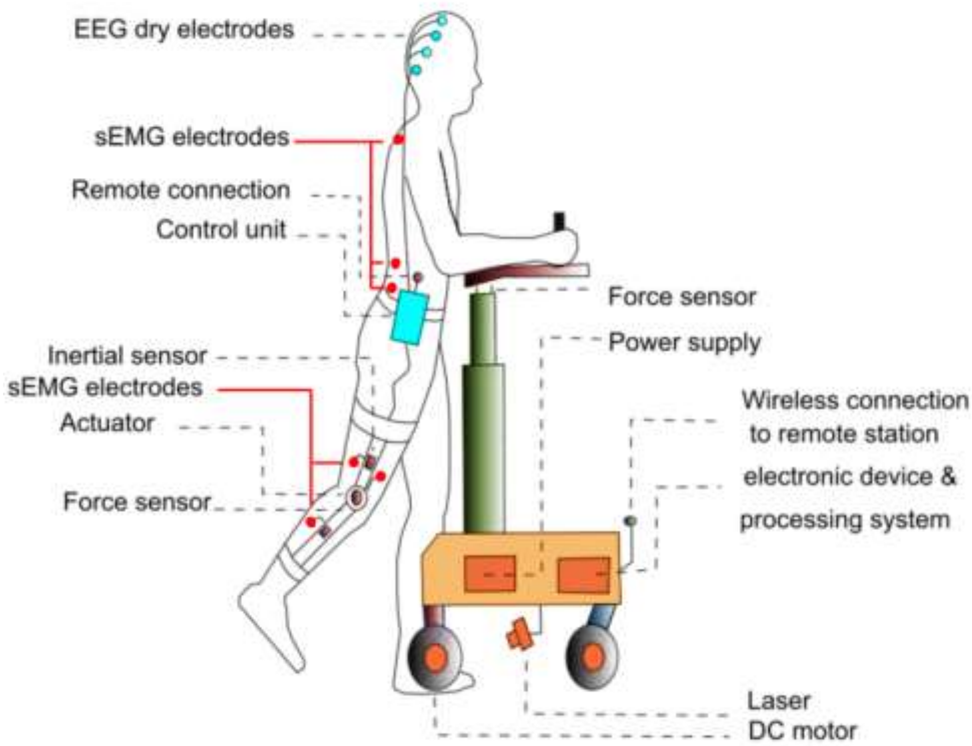

Figure 23. Rehabilitation robotic system composed of a robotic walker and a knee active exoskeleton. Source: (VILLA-PARRA et al., 2014).

The robotic walker is built with two forearm supports and a metallic rigid structure, which is suitable for people from 154 to 174 cm height, maintaining their upright posture. Moreover, its structure has two rear wheels driven by DC motors, and a front caster wheel. Apart from being a device to support the user's body weight, the robotic walker also provides information about the applied load on 3D force sensors located under each forearm support, which is one way to determine the user's motion intention. Nevertheless, the robotic walker device can adjust its movement to the patient speed through a laser range finder (LRF) sensor used to keep a fixed distance to the user's legs, employed for safety, avoiding collisions between the user legs and the walker. Additionally, there is a light detection and ranging (RP-LIDAR) sensor, located in front of the robotic walker, which detect obstacles, such as walls and people. Finally, optical shaft encoders and inertial sensor are used to provide the walker's position and orientation in real-time, respectively, whereas an embedded computer controls and processes the control tasks related to the device (JIMÉNEZ et al., 2018).

In addition to be commanded by the 3D force sensors, the rehabilitation robotic system can be either controlled by sEMG signals and/or brain (electroencephalography - EEG) signals, also providing information about the user's motor intention. Then, a controller sends the data to the actuator of the exoskeleton in order to execute the indicated task (VILLA-PARRA et al., 2014).

6.1. VOLUNTEERS

Two post-stroke volunteers were recruited in a rehabilitation institution (Center for Physical Rehabilitation of Espirito Santo – CREFES –, in Vila Velha/Brazil), following the inclusion criteria:

- The participant must be category 2 or greater than 2 of the Functional Ambulation Classification (FAC), described by Holden et al. (HOLDEN et al., 1984), which is used to evaluate the amount of human assistance, rather than devices, required for ambulation;
- Ability to stand erect and with elbows at around 90° when using robotic walker;
- Height among 154 and 174 cm, due to the limitation of the height adjust of the robotic walker;
- Cognitive and language skills sufficient to understand and follow the instructions of the experiment.

Additionally, the exclusion criteria were:

- Individuals who do not have independent gait;
- Have untreated cardiorespiratory impairment.

The study had approval of the UFES's Ethic Committee, and all volunteers signed the Free and Informed Consent Form (Number CAAE: 64797816.7.0000.5542). Figure 24 shows a post-stroke volunteer using the robotic walker.

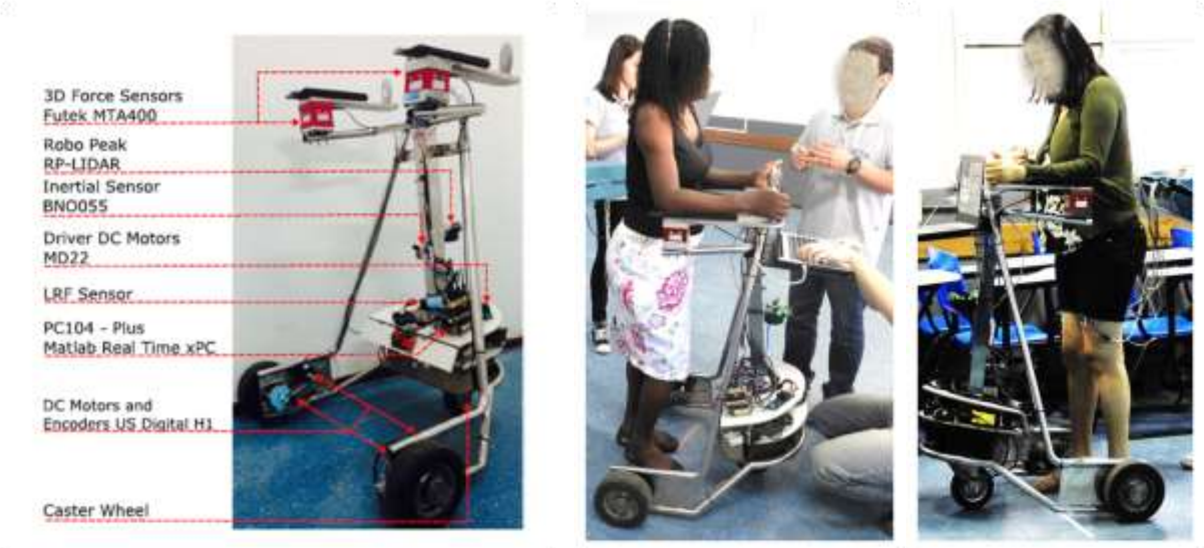

Figure 24. UFES's robotic walker (left) (JIMÉNEZ et al., 2018). The participant 1 receiving the orientations about the use of the robotic walker (middle). Participant 2 (right).

6.2. DATA COLLECTION AND ANALYSIS

In this study, we aimed to evaluate the symmetry and activation of the lower trunk and lower-limb musculature during the robotic walker's use. The procedure of acquisition and processing of sEMG signals was based on the recommendations of the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM, 2016), and such as suggested by (DE SÈZE et al., 2008; SWINNEN et al., 2012).

The sEMG channels were fixed to the following muscles on both sides: rectus femoris (RF) and erector spinae (ES) on two levels (T12 and L4). Also, a reference electrode was placed on the medial malleolus. On the other hand, a biaxial accelerometer was attached with adhesive tape on the L2 vertebra level of the subject. The sEMG and accelerometer data were recorded simultaneously using an acquisition equipment EMG 830C (EMG System do Brasil Ltda®) with sampling frequency of 1000 Hz.

All data analysis conducted in following case studies was similar to that described in Chapter 5.

6.3. EXPERIMENTS

Before starting the experiments, the International Physical Activity Questionnaire - Short Form (IPAQ-SF) was applied in order to assess the level of physical activity of each subject to classify them as physically active or inactive (LEE et al., 2011). According to the IPAQ-SF, both post-stroke subjects were classified as inactive and they reported performing activities only at home.

Initially, the volunteers walked at a comfortable speed on an 8-meter straight path and flat surface three times with no assistance. After each trial a necessary time of rest was allowed for each volunteer.

Prior to the use of the robotic walker, volunteers were advised on its operation and had a period of time for adaptation to the use of the device, and to outline the walker parameters. Thus, they walked under assistance of the robotic walker at a comfortable speed on an 8-meter straight path and flat surface three times.

At the end of all the experiments, the volunteers filled out the Modified Borg's scale (Table 111), indicating what was the effort level (ARVIN et al., 2015) during the use of the robotic walker. Also, they answered the System Usability Scale (SUS) questionnaire (BROOKE, 2013), which is a subjective evaluation of usability, in this case, about ease of use, ability to provide safety, confidence to walk and need for professional help to use the robotic walker (Annex E). Finally, more three extra questions were asked (Table 144).

Table 11. Modified Borg's scale.

Level	Effort
0	None
0.5	Very, very light
1	Very light
2	Light
3	Moderate
4	Slightly intense
5	Intense
6	-
7	Very intense
8	-
9	Very very intense
10	Maximum

6.4. RESULTS

6.4.1. Case #1

This is a 45 year old woman (164 cm height and 73 kg weight; Body Mass Index = 27 kg/m^2), who had one hemorrhagic stroke in the left hemisphere of the brain 5.5 years before the experiments, resulting from a fall during a convulsive crisis. As sequelae, she presented hemiparesis, with spasticity in the right lower-limb and right upper-limb and memory loss. Her spasticity in the knee joints was classified as level 1 on the Ashworth Modified Scale (Annex D), which indicates slight increase in muscle tone (BOHANNON; SMITH, 1987). The Functional Ambulation Category (FAC) was used to determine how much human assistance the patient requires when walking without use of devices. In order to use this scale, the participant walked a short distance, about 10 m, and was classified as category 4, which means to be an independent ambulator in level surface only.

Her speed during the gait with no assistance was 0.68 ± 0.05 m/s, and she presented a stance phase of 57.8 ± 3.1% and 59.1 ± 2.3% of gait cycle in the contralateral and ipsilateral limbs, respectively. Regarding the ratio contralateral/ipsilateral, the value obtained was 0.98 between the stance phases of each limb, which indicates high symmetry (considering that the closer the value is to 1.0 the more symmetrical are the sides).

After the gait assisted by the robotic walker, the participant 1 classified her level of effort as moderate in the Modified Borg's scale (score = 3). The speed using the walker was 3 times slower (0.22 ± 0.08 m/s) than the free gait and, consequently, the stance phase was prolonged in both limbs, being the toe-off of the contralateral limb in 68.7 ± 4.2%, and the ipsilateral in 69.4 ± 5.0%. In this case, also the ratio contralateral/ipsilateral for stance phase indicated high symmetry (0.99).

The muscle activation pattern of the participant 1, for each side in both gaits, is presented in Figure 25, as well as the percentage of the phase that each muscle kept activated. The muscle activation of the T12 and L4 presented a similar pattern, except by the later offset in the stance phase of the contralateral side during free gait. From the measurements, it was possible to observe that, for T12 and L4, only the contralateral side in the assisted gait had an activation, which began and finished in the stance phase, differently of others, which finished in the swing phase, such as observed in healthy subjects (CECCATO et al., 2009; KARTHIKBABU et al., 2012).

In the case of the RF muscle, the contralateral and ipsilateral in the free gait presented abnormal onset/offset timing in the central region of the gait cycle. The contralateral RF was activated after toe-off, whereas the ipsilateral RF finished its activation in the early, in the stance phase. Using the robotic walker, both contralateral and ipsilateral RF showed the activation around the toe-off, the onset occurred in stance phase and the offset in the swing phase.

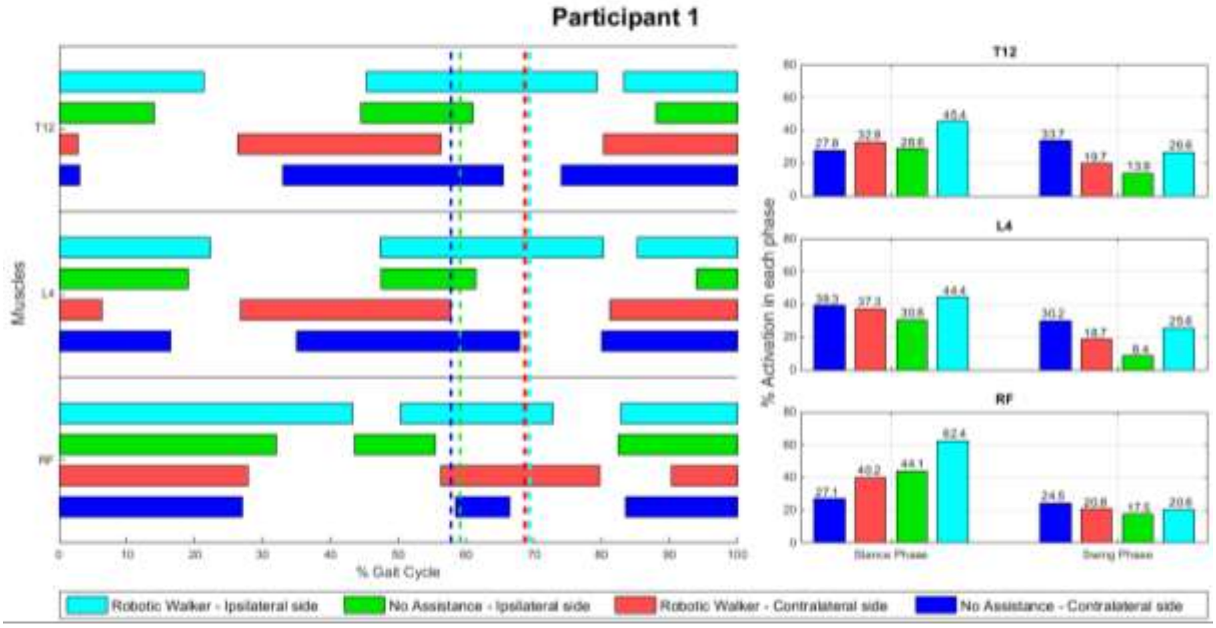

Figure 25. Muscle activation pattern of the participant 1 during walking with no assistance and with robotic assistance, for each side and each muscle (left). Percentage of activation of each muscle in both stance and swing phases (right). T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

In Table 12 the ratio contralateral/ipsilateral for the duration of activation in each phase is presented, which is calculated from the values of Figure 25. For this participant, the use of the robotic walker improved the symmetry of duration of activation in the swing phase for all muscles. On the other hand, there were changes in the stance phase, although they are heterogeneous.

Table 12. Ratio contralateral/ipsilateral for duration of activation in stance and swing phase for the participant 1, which was calculated to analyze the symmetry between contralateral and ipsilateral sides.

		Stance Phase		Swing Phase	
		No assistance	Robotic Walker	No assistance	Robotic Walker
Part. 1	**T12**	0.97	0.72	2.42	0.74
	L4	1.28	0.84	3.60	0.73
	RF	0.61	0.64	1.40	1.01

T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

Finally, the total SUS score of the participant 1 was 77.5 (Table 144), which means this participant considered the robotic walker is usable, and, according to the answers of the extra questions, she felt she had the control over handling, her interaction with the robotic walker was very easy to understand, and she got used quickly to its use.

6.4.2. Case #2

This is a 48 year old woman (165 cm height and 68 kg weight; Body Mass Index = 25 kg/m^2), with a history of arterial hypertension and diabetes. She had one ischemic stroke in the left hemisphere of the brain 8 months before the experiments. As sequelae, she presented hemiparesis, with spasticity in the right lower-limb and right upper-limb. Her spasticity in the knee joint was classified as level 2 on the Ashworth Modified Scale, which indicates more marked increase in muscle tone through most of the range of motion, but the affected parts are easily moved. She was classified as ambulator dependent for supervision, or category 3 in the FAC.

She had more difficulty walking and holding the walker handle, due to her level of spasticity and the stroke has been more recent than participant 1. When walking without assistance, her gait speed was 0.36 ± 0.06 m/s, and using the robotic walker, her speed reduced to 0.18 ± 0.03 m/s.

The stance phase during the free gait lasted 55.5 ± 5.2% of the gait cycle for the contralateral limb, whereas in the ipsilateral limb it lasted 63.1 ± 3.8 %. Therefore, the ratio contralateral/ipsilateral was of 0.88. During the assisted gait, the toe-off of the contralateral limb was 65.8 ± 3.3%, and of the ipsilateral was 70.3 ± 4.0%, resulting in a small increase in symmetry (0.94). The participant 2 also evaluate her level of effort during the gait assisted by the robotic walker as moderate (score = 3).

The parameters related to muscle activation of the participant 2 are showed in Figure 26. The T12 and L4 patterns were relatively similar to each other, however, the contralateral side in the free gait was more activated during swing phase than in the assisted gait, and the offset ipsilateral was very close to the toe-off. Regarding the swing phase, the ipsilateral side showed an earlier onset than the contralateral side, mainly in gait without assistance.

In all the cases, the RF activation near to toe-off was longer than the healthy gait. In the assisted gait, the ipsilateral RF had two offsets in the stance phase, i.e., the second offset occurred earlier than the expected (PERRY; BURNFIELD, 2010; WARD et al., 2018).

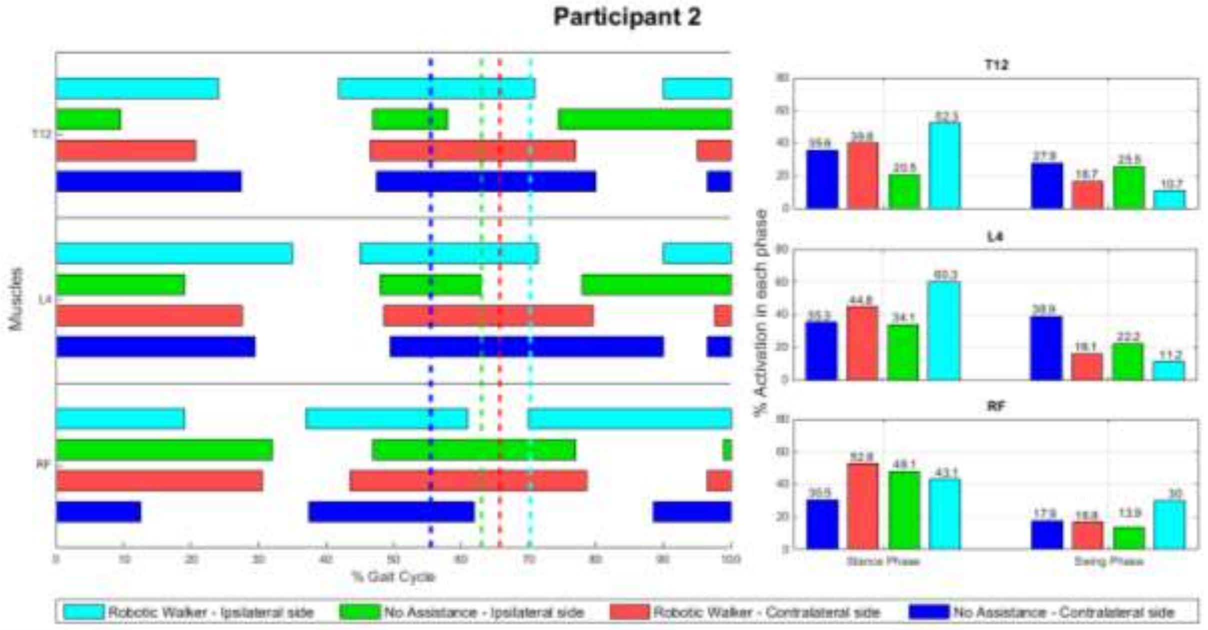

Figure 26. Muscle activation pattern of the participant 2 during walking with no assistance and with robotic walker assistance, for each side and each muscle (left). Percentage of activation of each muscle in both stance and swing phases (right). T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

The ratio contralateral/ipsilateral of the duration of activation was heterogeneous in both phases for the participant 2 (Table 13). There was an increased symmetry using the robotic walker in T12 and RF, during stance phase, and in L4 during swing phase. However, there was a decreased symmetry for the others activations.

Table 13. Ratio contralateral/ipsilateral for duration of activation in stance and swing phase for the participant 2, which was calculated to analyze the symmetry between contralateral and ipsilateral sides.

		Stance Phase		**Swing Phase**	
		No assistance	Robotic Walker	No assistance	Robotic Walker
Part. 2	**T12**	1.74	0.76	1.09	1.56
	L4	1.04	0.74	1.75	1.44
	RF	0.63	1.23	1.29	0.56

T12 and L4 are the erector spinae levels analyzed; RF: rectus femoris.

The participant 2 then assessed the robotic walker usability, giving a total score of 70.0, which is an above-average value, i.e., the walker was considered usable. Analyzing the extra questions, she agreed with the first statement and strongly agreed with the others, as shown in Table 144, which represents a positive result for her interaction with the robotic walker.

Table 14. Scores given to each item in the questionnaires.

	System Usability Scale (SUS) statements*											Extra questions*		
	#1	**#2**	**#3**	**#4**	**#5**	**#6**	**#7**	**#8**	**#9**	**#10**	**Total****	**#a**	**#b**	**#c**
Part. 1	5	1	4	4	5	2	5	1	5	2	**77.5**	4	5	5
Part. 2	4	1	4	4	5	2	4	1	4	4	**70.0**	4	4	5

#a - I felt I have control over the handling of the robotic walker.
#b - I felt the interaction with the robotic walker was easy to understand.
#c - I got used to the use of the robotic walker.
* The statements or questions had five options, being (1) strongly disagree, (2) disagree, (3) neutral, (4) agree, and (5) strongly agree.
** The total score was calculated following the orientations of (BROOKE, 2013).

6.5. CONCLUSIONS

Two post-stroke individuals performed experiments using the robotic walker, whose characteristics were very different for each other, since the type of stroke, time after stroke, etc. The use of the robotic walker reduced the gait speed in both cases, increasing the stance phase in the contralateral and ipsilateral limbs, however, their symmetry increased for the participant 2.

Both participants did not present big changes during the use of the robotic walker, however, for the participant 1, it improved the symmetry of duration of activation in the swing phase for all muscles, and both contralateral and ipsilateral RF showed activation closer to the healthy gait.

Regarding the opinion of the participants about the usability of the robotic walker, the results were considered satisfactory, and both participants had a good interaction and adaptation to the device.

7. ELECTRONIC DEVICE FOR POSITION SENSING AND SYNCHRONIZATION OF BIOLOGICAL DATA

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** Filing the patent application – INIT/UFES*

7.1. SUMMARY

The invention described here is referred as an Electronic Device for Position Sensing and Biological Data Synchronization, composed of two modules: Position and Synchronism Sensor (SPS) and Converter and Synchronizer (CS). The SPS module consists of an IMU sensor (Inertial Measurement Unit), an AVR microcontroller for internal processing, and has wireless communication through Bluetooth protocol. The CS module is a digital signal converter for two independent analog signal outputs, uses Bluetooth protocol for communication with the SPS module and serial communication through a micro USB connector.

The SPS acts as a biomechanical signal transducer in digital signals, which can be used to synchronize the movement performed by the patient during the rehabilitation exercises to his/her avatar in a virtual reality (VR) environment. This module can be used directly through Bluetooth communication when used for interaction with the VR. However, to synchronize the lower/upper limbs movements to electroencephalography (EEG) and surface electromyography (sEMG) signals, it is necessary to use the CS module, which has biomechanical and bioelectrical signals synchronization function. Thus, it is possible to obtain data from different amplifiers with a common signal, facilitating data processing.

7.2. BACKGROUND OF THE INVENTION

Patients with lower limbs motor impairment may benefit from physical rehabilitation, in which the most targeted goal is the recovery of individual's independence in basic tasks (walking, bathing, doing household chores). Functional recovery of motor deficits in neurological patients may require a considerable amount of movement repetition to induce changes in neuroplasticity. The gold standard method in rehabilitation, aerobic exercises, in addition to inducing a high number of repetitions, has the potential to promote improvements in the circulatory, respiratory and muscular systems. Improved blood supply and uptake of oxygen by tissues, increase and maintenance of active joint amplitude, and preservation of muscle tissue are some benefits that help to preserve the patients' health and contribute to their recovery and rehabilitation.

As an example, walking and cycling training are useful because they are repetitive tasks, easy to perform, activate various muscle groups, promote improved blood circulation, respiratory capacity and maintenance of muscle tissue.

Because it is a repetitive and monotonous training, the patient may present lack of attention, demotivation and even withdrawal from therapy. It is important that patients play an active role in their rehabilitation process, since those who are more motivated do have a better recovery. As a way to increase the patient motivation, Serious Games (SG) in a Virtual Reality (VR) environment can be an alternative, as they provide a playful form of rehabilitation, providing immersive biofeedback, and also provide a cognitive rehabilitation, since the patient needs to pay attention to the goals of the game.

In this way, VR can provide the patients with a varied and enjoyable environment that implies their motivation to practice the movements needed for rehabilitation over long periods of time. The possibility of adding to these patients simultaneous feedback, knowledge of performance and the results achieved in the rehabilitation process are important for learning as well as motivation, and will directly influence their recovery. Studies indicate that patient motivation is a highly important factor for the end result of the therapeutic process.

Additionally, incorporating information on biological signals through EEG and sEMG in VR-based games aims to provide the rehabilitation professional with information to evaluate the evolution of the patients and to establish individualized and more precise goals aimed at their rehabilitation. It is important that the biological signals are synchronized with the biomechanical signals so that the phase of movement in which the muscle is contracted or which region of the brain is being activated can be determined. These data can be compared to the pattern of people without motor impairment, already described in the scientific literature.

7.3. DETAILED DESCRIPTION OF THE INVENTION

The invention described here aims to provide accurate synchronization of the positioning angles either in apparatuses as monocycle or bicycle or even positioned directly on limbs or other body regions of the patient. It use can be used in physical rehabilitation research protocols, as well as in physiotherapy and occupational therapy protocols, and para-sport training.

The present invention proposes two modules, which, when used together, allow the synchronization between the patient´s movement and his/her EEG/sEMG signals. Also, while using only the Position and Synchronization Sensor (SPS), the patient´s movements can be used directly on computer or smartphone applications.

The apparatus described here allows the acquisition of positioning by transducing the signals obtained by the inertial sensors into digital signals, which are transmitted using Bluetooth protocol. The SPS module is responsible for this transduction and data transmission. Data can be sent to computers or other portable devices, such as Android-based devices.

The data can also be received by the CS module, where it is converted into an 8-bit analogue resolution signal. After this conversion, the signal is provided, isolated, through two channels. These channels can be configured via a mini-switch to use the 0-5 V DC of the module itself, or to use the voltage of the EEG / sEMG equipment. The information processing is performed by AVR microcontroller, which manages two digital potentiometers for generating the output signals. This signal can be configured to have power independent of the device, that is, be supplied by the devices themselves where it will be used, or, if necessary, be fed by the internal circuit. Unlike the SPS module, which can be used separately, this module only has its functionalities when used in conjunction with the SPS module. It is a tool with great utility when it is desired to relate inertial information to other signals, such as sEMG and EEG. Due to the fact that it has two isolated outputs providing a common signal, it can be used with a reference signal to perform the synchronization between the two devices, facilitating subsequent data processing.

The sensor can also be configured via numerical commands to provide information from temperature, gyroscope and accelerometer data without any processing. The data sampling frequency can also be changed, set by default to 100 Hz, depending on its signal sending mode, configured as continuous or request dependent.

The invention described here proposes, through the CS module, a new functional concept that resides in the analogous synchronization of EEG and sEMG signals to physical events. Such synchronization is performed by generating a common signal sent to both equipment and voltage levels isolated and proper to each one.

The implementation of this invention allows the obtaining of signals through different equipment, with great ease for synchronization without complex processing. In addition, the devices constructed have the characteristic of being easy to use and low cost.

7.4. UTILITIES

The SPS module can be used on a unicycle crank to identify its angles during pedaling and thus reproduce (faithfully) the movement in a SG. That is, the patient will pedal a unicycle containing the module and the patient will have this same movement being done in the virtual environment by an avatar. It also identifies the propulsive phase (0° to 180°), where the rider applies the greatest force on the pedal, the recovery phase (180° to 360°) and the rotation of the crank.

In addition, the SPS module can send information about the positioning of the crankcase to the CS module. Thus, the CS module, connected to a biological signal acquisition device, will synchronize both signals. This synchronization allows biological signals to be analyzed offline. The pattern of muscle activation (acquired through sEMG) and analysis of patterns of brain signals (acquired through EEG) can be used to evaluate the progression of patient recovery, among others.

Another utility of SPS is its use on the patient's body. With SPS disposed on the ankle or pelvis, it is possible to identify support phases and gait balance. Specifically, with the positioning on the pelvis, it is possible to evaluate the gait symmetry, a variable widely used for the evaluation of walking improvement in some patients, such as patients with post-stroke hemiparesis. This information on gait phases is important even if one wants to analyze the muscular activation pattern in this task, in whose case the CS would be used to synchronize IMU signals with biological signals.

The acquired data, in addition to be used for the physical evaluation of the patients, can be used to reproduce the movements of the individual in a serious game. For this, the patient can use a treadmill or walk on the ground, depending on how the AV is presented to him/her. It can be displayed in front of him/her using a projector, displayed on a screen, or use a Head Mounted Display, as examples. The SPS can also be positioned on the arm in order to reproduce the movement performed by the patient in the virtual environment.

7.5. CLAIMS

1) Position and Synchronism Sensor (SPS) comprising:
   a) an IMU sensor "Inertial Measurement Unit"
   b) an AVR microcontroller for internal processing
   c) a micro USB connector (12), which is used to charge the internal battery
   d) wireless communication through Bluetooth protocol, characterized by acting as a transducer of biomechanical signals in digital signals, which can be used to synchronize the movement performed by the patient during

the rehabilitation exercises to his/her avatar in a virtual reality environment.

2) Converter and synchronizer (CS) comprising:

   a) a digital signal converter for two independent analog signal outputs (22, 23)

   b) wireless communication via Bluetooth protocol for communication with the SPS module

   c) a charging connector (21)

   d) a connector for serial communication via USB, characterized by having the functionality of synchronizing two biological signal monitoring apparatuses, such as Electroencephalography (EEG) and surface Electromyography (sEMG), among others, to an inertial signal from SPS (according to claim 1) through the two analog outputs (22, 23).

3) Fastening bracket, comprising:

   a) holder for fixing to the crank, characterized by having the ability to position the SPS module to the bicycle and monocycle crank used for physical rehabilitation;

   b) fastening bracket to the body, characterized by having the ability to position the SPS module to the ankle, arm or hip of the patient.

7.6. FIGURES

Figure 1 shows a drawing of the SPS device.

Figure 2 is a presentation of the CS module.

Figure 3 is the fastening bracket of the SPS device for positioning the monocycle and/or bicycle crankcase.

Figure 4 is the fastening bracket of the SPS device for positioning, using elastic band, adjacent to the subject's body (arm, leg or back)

Figure 5 is a representation of the SPS device mounted adjacent to the tape holder.

Figure 6 is a representation of the SPS device mounted adjacent the crank support.

Figure 1 shows the on/off button (11), the micro USB connector (12), which is used for charging the internal battery. The region of the cover represented by the number 13 indicates the location where the operation LED flashes. The lower base of the device (14) and its side (15) are used as reference for positioning.

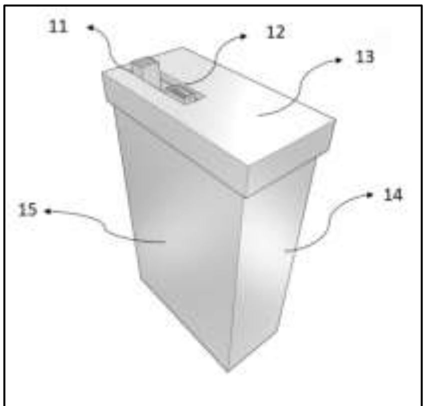

Fig. 1

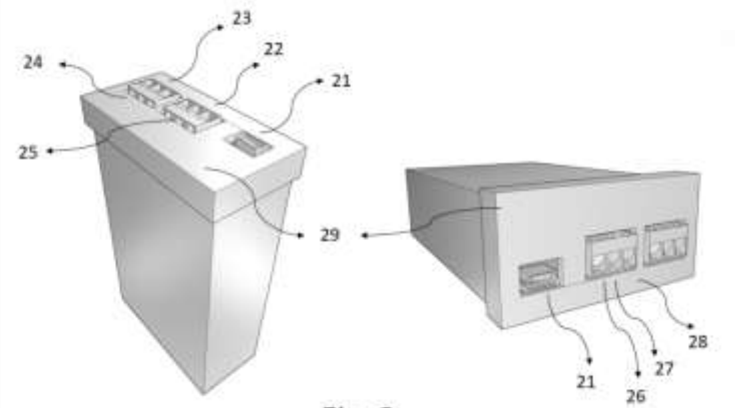

Fig. 2

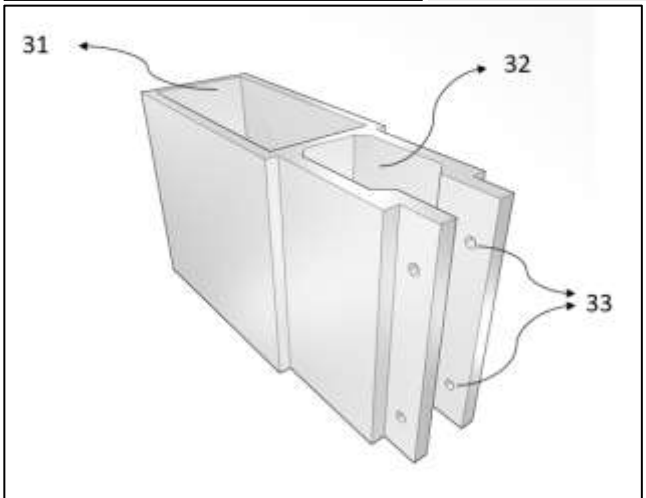

Fig. 3

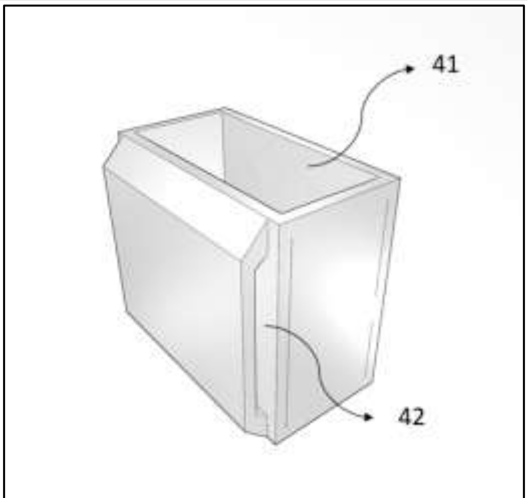

Fig. 4

Figure 2 shows the CS signal converter module, the charging connector (21) can also be used for serial communication via USB. The analog outputs (22, 23) are used to connect the biological signal amplifiers, such as EEG and sEMG. Due to the wide variety of cables available for each type of amplifier, a generic connector was used, on which the grip is done by tightening the screws (24, 25). LED function indicator is represented by the number 28. Each of the output channels (22, 23) is composed of a 3-way connector: ground (GND) (26), analogic signal (SIGNAL OUT) (27) and power (VCC) (28), in which the devices that will amplify the biological signals are connected. The power input can be made either external (26, 28) or directly by the CS, being the choice made through the programming parameters. When the devices used have two-way channels only, the connectors represented by numbers 26 and 27 are used, and, in this case, the power input is done by the internal circuit of the CS.

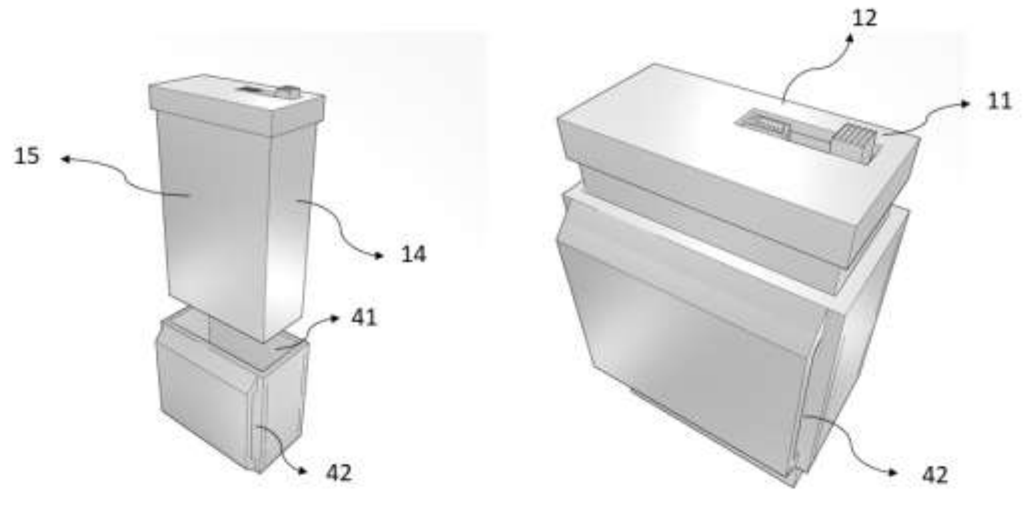

Fig. 5

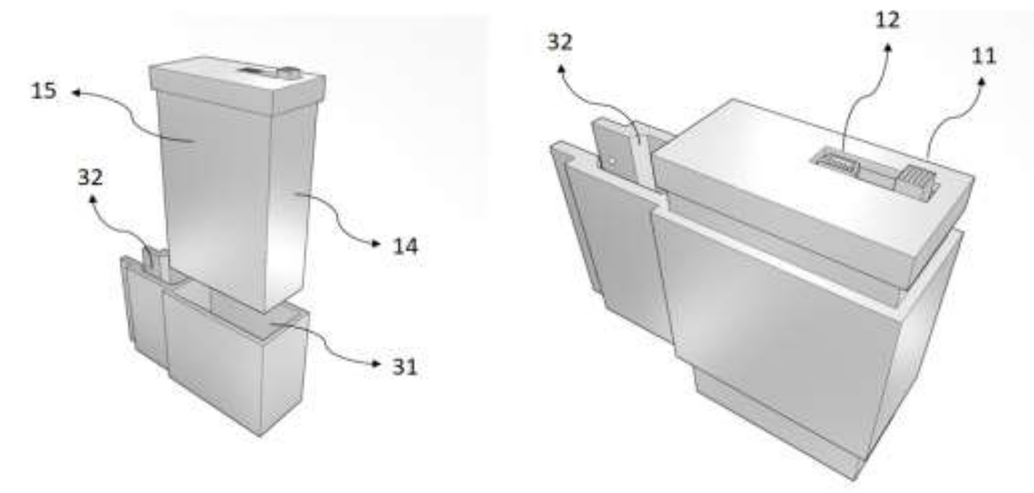

Fig. 6

The fastening bracket (Figure 3) is positioned to the crank through the recess (32) and is secured by means of M3 screws 30 mm long, inserted into the holes represented by the numeral 33. The SPS device is placed in the part represented by the number 31, being fixed by pressure without any other form of locking, and the area represented by the number 15 is positioned to the right side, regarding the positive direction of the linear movement.

The body attachment support (Figure 4) is used to position the SPS device (Figure 1) on the ankle, wrist or back of the subject.

8. FINAL CONSIDERATIONS

Trunk muscle functions are essential to perform a gait with adequate energy expenditure, posture, and stability. ES muscle was sequentially activated from C7 to L4 in both healthy and stroke group during different gait modalities, presenting, in all cases, two periods of activation in the cycle gait. The results obtained in this study showed that there is no influence of arm swing in the ES activation, and the conventional walker did not alter their muscle patterns. However, during gait on a treadmill, changes can occur in the activation of trunk and lower-limb muscles.

Neuromuscular fatigue can be detected through of the shift of the median frequency (MDF) to lower values, in both isometric and dynamic contractions. In fact, it occurred gradually during all the movements carried in this research in non-strenuous exercises. During gait on treadmill, all of three lower-limb muscles presented a decreased MDF, whereas only L4 of three ES muscle levels had a reduction in MDF.

Post-stroke subjects have the trunk function altered bilaterally, but more preserved than lower-limb muscle functions. It was observed in this study that, in the stroke group, ES muscle presented a similar pattern to the healthy group, however, its activation near the toe-off was longer, probably due to a longer double support before the contralateral limb swing. On the other hand, the use of walker reduced significantly the excessive BF activation in the contralateral side of the post-stroke subjects.

The hemiparesis causes a remarked asymmetry in the gait parameters in post-stroke individuals. In fact, it was observed in our research that both contralateral and ipsilateral stance phases had significant changes when compared to healthy individuals, which were not modified through the use of the conventional walker. Regarding the muscle activity, the contralateral ES muscle had longer activation (near the toe-off) than the ipsilateral side in both free and assisted gait.

Comparing the activation in each phase of both groups, only the ipsilateral T12 and contralateral RF activations did not present statistically significant difference in the stance phase of the free gait, and all the observed changes indicated a longer duration of activation of the stroke group, except for ipsilateral T12 activation in the swing phase during free gait. The ratio contralateral/ipsilateral became closer to 1.0 with the walker assistance, being that the walker reduced the asymmetry at 5 out of 6 measurements.

About neuromuscular fatigue in stroke group, it was not possible to detect reduced MDF, possibly due to the fact the gait task was too light.

Finally, two post-stroke individuals performed experiments using the robotic walker developed at UFES. The gait speed was reduced during its use, however, one of the participants had her stance phase symmetry increased, and the other participant had improvement in her symmetry of duration of activation in the swing phase for all muscles. In addition, both contralateral and ipsilateral RF showed activation closer to the healthy gait. According their results for SUS (77.5 and 70.0), they considered that they had a good interaction and ease adaptation to the robotic walker.

8.1. FUTURE WORKS

- Test trunk muscle sEMG signals as input to control an exoskeleton;
- Perform preliminary experiments using the robotic rehabilitation system in healthy and post-stroke subjects;
- Elaborate and apply a rehabilitation protocol using the robotic rehabilitation system in post-stroke subjects during therapy sessions;
- Analyze muscle activation and fatigue during therapy sessions with robotic rehabilitation system and verify the rehabilitation progress.

8.2. PUBLICATIONS DURING RESEARCH

8.2.1. Journals

Delisle-Rodriguez, D; Cardoso, VF; Gurve, D; Loterio, FA; Romero-Laiseca, MA; Krishnan, S; Bastos, TF . System based on subject-specific bands to recognize pedaling motor imagery: Towards a BCI for lower-limb rehabilitation. **Journal of Neural Engineering**, v. 1, p. 1-29, 2019.

Loterio. FA; Valadão. CT; Cardoso. VF; Pomer-Escher. A; Bastos. TF & Frizera-Neto. A. Adaptation of a smart walker for stroke individuals: a study on sEMG and accelerometer signals. **Research on Biomedical Engineering**. 33(4). 293-300. (2017). Epub November 09. 2017.https://dx.doi.org/10.1590/2446-4740.01717

Pomer-Escher. A; Loterio. FA; Longo. BB; Glasgio. G; Bastos. T. Evaluación de La Sensación de Presencia en un Ambiente Virtual para Neurorehabilitación. **Cognitive Area Networks.** v. 3. p. 13-18. 2016.

Cardoso. V; Valencia. N; Loterio. FA; Frizera-Neto. A; Bastos. T. Análisis de EMGs en Ambiente de Realidad Virtual para Rehabilitación de Miembros Superiores de Pacientes Post-Ictus. **Cognitive Area Networks**. v. 3. p. 39-44. 2016.

Valadão. CT; Loterio. FA; Cardoso. V; Frizera-Neto. A; Carelli. R; Bastos. T. Robotics as a Tool for Physiotherapy and Rehabilitation Sessions. **IFAC-PapersOnLine**. v. 48. n. 19. 2015.

8.2.2. Articles submitted to Journals

Loterio, FA; Pomer-Escher, A; Cardoso, VF; Valadão, CT; Bastos-Filho, TF and Frizera-Neto, A. Effect of Different Modalities of Gait on Erector Spinae and Lower-Limb Muscles Activation Pattern. **Journal of Electromyography and Kinesiology**. Submitted: February, 2019.

Loterio, FA; Pomer-Escher, A; Cardoso, VF; Valadão, CT; Bastos-Filho, TF and Frizera-Neto, A.. Identification of Neuromuscular Fatigue during Gait on Treadmill and Isometric Exercises Through Short-Time Fast Fourier Transform. **Journal of Electromyography and Kinesiology**. Submitted: February, 2019.

Loterio, FA; Pomer-Escher, A; Valadão, CT; Cardoso, VF; Bastos-Filho, TF; and Frizera-Neto A.. Electromyography Analysis of Trunk and Lower-Limb Muscles of Post-Stroke Individuals during Free and Walker-Assisted Gait. **Disability and Rehabilitation.** Submitted: February, 2019.

8.2.3. Patents

Pomer-Escher, A; Longo, BB; Loterio, FA; Cardoso, VF and Bastos-Filho, TF. Electronic Device for Position Sensing and Synchronization of Biological Data. Filing the patent application – **INIT/UFES**, 2019.

8.2.4. Full papers in conference proceedings

Loterio, FA; Cardoso, VF; Pomer-Escher, A; Bastos-Filho, TF; Frizera-Neto, A. Krishnan, S. Identification of kinematic parameters of gait using an accelerometer. **Anais do Congresso Brasileiro de Engenharia Biomédica (CBEB2018)**, Búzios, 2018.

Cardoso, VF; Pomer-Escher, A; Longo, BB; Loterio, FA; Nascimento, S; Delisle, D; Frizera-Neto, A; Bastos-Filho TF. Neurorehabilitation Platform Based on EEG, sEMG and Virtual Reality Using Robotic Monocycle. **Anais do Congresso Brasileiro de Engenharia Biomédica (CBEB2018)**, Búzios, 2018.

Loterio. FA ; Cardoso. VF; Valadão. CT; Frizera-Neto. A; Bastos. TF. Detecção de Fadiga Muscular através de Análise de Imagem Térmica e Eletromiografia de Superfície. **Anais do Simpósio Brasileiro de Automação Inteligente (SBAI2017)**. Porto Alegre. 2017.

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APPENDIX A - FREE AND INFORMED CONSENT FORM

TERMO DE CONSENTIMENTO LIVRE E ESCLARECIDO

O(A) Sr.(a) ____ foi convidado (a) a participar da pesquisa intitulada Desenvolvimento de dispositivos de tecnologia assistiva e reabilitação baseados em realidade virtual e sinais biológicos (coleta de sinais mioelétricos), sob a responsabilidade do Prof. Dr. Teodiano Freire Bastos-Filho.

JUSTIFICATIVA

A quantidade de pessoas com deficiência ou com dificuldade de realizar atividades do dia-a-dia (por exemplo: andar. subir escadas. vestir-se. escrever e outras coisas) vem aumentando no mundo todo. Isso pode ser causado pelo envelhecimento, doenças como derrame ou acidentes de trânsito. Assim, é importante pensar no desenvolvimento de equipamentos, robôs e jogos de computador usando tecnologias mais modernas para ajudar no tratamento e no dia-a-dia dessas pessoas.

OBJETIVO(S) DA PESQUISA

Desenvolver e avaliar equipamentos, robôs e jogos de computador que possam ajudar no tratamento e no dia-a-dia das pessoas com deficiência e dificuldades de realizar atividades diárias.

PROCEDIMENTOS

Você responderá um questionário, com perguntas sobre: seus dados pessoais, socioeconômicos e sobre a doença e suas sequelas. Para coletar sinais dos músculos, os pesquisadores irão limpar a pele em cima do músculo que vamos pegar o sinal com álcool 70%, raspar os pelos no local que serão colados os adesivos, colar dois adesivos pequenos (que possuem um anel metálico e gel no meio) em cima da pele na direção que fica o músculo e outro em um local que não tenha músculo. Uma pulseira que lê sinais musculares e um óculos de realidade virtual podem ser usados também.

Durante os testes o pesquisador irá falar e mostrar os movimentos que o participante irá fazer, por exemplo: esticar e dobrar o joelho, levantar e abaixar a perna, esticar e dobrar o cotovelo, levantar e abaixar o braço, rodar o braço, sentar e levantar, mexer o tronco para direita e para esquerda, caminhar, pedalar. O pesquisador irá mostrar quantas vezes for preciso. O participante pode fazer o movimento antes de começar até que se sinta seguro. O participante poderá usar um ambiente de realidade virtual na tela do computador ou um óculos de realidade virtual, cada jogo ou ambiente de realidade virtual será usado de 3 a 5 vezes, com descanso entre eles. No final você responderá questionários para avaliar o equipamento e o ambiente de realidade virtual.

DURAÇÃO E LOCAL DA PESQUISA

Os testes serão feitos no Núcleo de Tecnologia Assistiva, que fica no Departamento de Engenharia Elétrica da Universidade Federal do Espírito Santo – campus Goiabeiras e/ou no local de seu tratamento (centro de reabilitação. hospital. ONG), ou em sua casa, dependendo do tipo de teste. Cada dia que você participar da pesquisa vai demorar no máximo 1 hora e 30 minutos. Nesse tempo vamos fazer o seguinte: colocar os equipamentos, explicar e demonstrar os movimentos, usar o equipamento, descansar e responder os questionários.

RISCOS E DESCONFORTOS

Você pode se sentir cansado durante os movimentos. Por causa disso, haverá pausas para descanso. Se você precisar de uma pausa maior para descansar, será dado mais tempo. Você sempre será acompanhado de um profissional da saúde durante toda a pesquisa.

BENEFÍCIOS

Nós esperamos que os resultados destes testes ajudem a desenvolver equipamentos e avaliar equipamentos, robôs e jogos de computador que possam ajudar no tratamento e no dia-a-dia de pessoas com deficiência e de pessoas com dificuldades de realizar suas atividades diárias.

ACOMPANHAMENTO E ASSISTÊNCIA

Durante toda a pesquisa você poderá se comunicar com os pesquisadores, informando quaisquer problemas ou dificuldades com o uso do equipamento. Os pesquisadores asseguram a assistência imediata e integral por quaisquer danos decorrentes da pesquisa.

GARANTIA DE RECUSA EM PARTICIPAR DA PESQUISA E/OU RETIRADA DE CONSENTIMENTO

Você não é obrigado(a) a participar da pesquisa, podendo deixar de participar a qualquer momento, sem que haja penalidades ou prejuízos. Caso você não queira mais participar, os pesquisadores não entrarão mais em contato com você.

GARANTIA DE MANUTEÇÃO DO SIGILO E PRIVACIDADE

Você terá sua identidade e suas imagens preservadas durante todas as fases da pesquisa, inclusive após publicação.

GARANTIA DE RESSARCIMENTO FINANCEIRO

Todas as despesas relativas ao seu deslocamento dos seus familiares, caso seja necessário, e outras despesas que possam surgir com sua participação nesta pesquisa serão cobertas pelos pesquisadores.

GARANTIA DE INDENIZAÇÃO

Diante de danos que possam acontecer por causa da pesquisa, você será indenizado pelos pesquisadores.

ESCLARECIMENTO DE DÚVIDAS

Em caso de dúvidas ou problema sobre a pesquisa, você pode entrar em contato com o pesquisador TEODIANO FREIRE BASTOS-FILHO no telefone (27) 4009-2077 ou endereço Av. Fernando Ferrari. 514. Goiabeiras. CEP: 29075910 - Vitória. ES - Brasil. Em caso de denúncias e/ou problemas na pesquisa, você também pode entrar em contato com o Comitê de Ética em Pesquisa da Universidade Federal do Espírito Santo - Goiabeiras (CEP/Goiabeiras/UFES), através do telefone (27) 3145-9820, e-mail cep.goiabeiras@gmail.com ou correio:

> Comitê de Ética em Pesquisa com Seres Humanos. UFES/Campus Goiabeiras. Prédio Administrativo do Centro de Ciências Humanas e Naturais. Sala sete. Campus Universitário de Goiabeiras. Av. Fernando Ferrari. 514. Vitória – ES. 29075-910

O CEP/Goiabeiras/UFES tem a função de analisar projetos de pesquisa para proteger os participantes dentro de padrões éticos nacionais e internacionais.

Declaro que fui verbalmente informado e esclarecido sobre o presente documento, entendendo todos os termos acima expostos, e que voluntariamente aceito participar deste estudo. Também declaro ter recebido uma via deste Termo de Consentimento Livre e Esclarecido, de igual teor. assinada pelo(a) pesquisador(a) principal ou seu representante e também por mim ou meu responsável legal, rubricada em todas as páginas.

______________________________ ____/____/______.

____

Participante da pesquisa/Responsável legal

Na qualidade de pesquisador responsável pela pesquisa Desenvolvimento de dispositivos de tecnologia assistiva e reabilitação baseados em realidade virtual e sinais biológicos, eu, Teodiano Freire Bastos-Filho, declaro ter cumprido as exigências do(s) item(s) IV.3 e IV.4 (se pertinente), da Resolução CNS 466/12, a qual estabelece diretrizes e normas regulamentadoras de pesquisas envolvendo seres humanos.

____

Pesquisador

ANNEX A – PATENT DEPOSIT

UNIVERSIDADE FEDERAL DO ESPÍRITO SANTO
Pró-Reitoria de Pesquisa e Pós-Graduação
Instituto de Inovação Tecnológica

DECLARAÇÃO

Declaro, para os devidos fins, que foi protocolado junto a este INIT – Instituto de Inovação Tecnológica da UFES, na data de 08 de janeiro de 2019, a solicitação de pedido de patente com o título "**Dispositivo Eletrônico para Leitura de Posição e Sincronia de Dados Biológicos**", tendo a UFES – Universidade Federal do Espírito Santo como titular e seus inventores, Alexandre Geraldo Pomer-Escher, Berthil Borges Longo, Flávia Aparecida Loterio, Vivianne Flávia Cardoso e Teodiano Freire Bastos Filho.

Vitória/ES, 16 de janeiro de 2019.

José Carlos Fundão Farias
Administrativo – INIT/UFES

Jose Carlos [illegible]
[illegible]

ANNEX B - FUNCTIONAL AMBULATION CATEGORY (FAC)

This scale was developed at Massachusetts General Hospital, first described by (HOLDEN et al., 1984), the Functional Ambulation Category (FAC) assess the functional walking ability, being divided into 6 categories. It involves the determination of how much human care the patient requires when walking, without the use of devices (HOLDEN et al., 1984). The FAC does not assess the resistance because the patient is evaluated in a gait of about 10 steps. It can be used with, but not limited to, stroke individuals.

Category	Definition
0 – Nonfunctional ambulation	Subject cannot ambulate, ambulates in parallel bars only. or requires supervision or physical assistance from more than one person to ambulate safely outside of parallel bars
1 – Ambulator Dependent for Physical Assistance Level II	Subject requires manual contacts of no more than one person during ambulation on level surfaces to prevent falling. Manual contacts are continuous and necessary to support body weight as well as maintain balance and/or assist coordination.
2 – Ambulator Dependent for Physical Assistance Level I	Subject requires manual contact of no more than one person during ambulation on level surfaces to prevent falling. Manual contact consists of continuous or intermittent light touch to assist balance or coordination.
3 – Ambulator Dependent for Supervision	Subject can physically ambulate on level surfaces without manual contact of another person but for safety requires standby guarding on no more than one person because of poor judgment, questionable cardiac status, or the need for verbal cuing to complete the task.
4 – Ambulator Independent Level Surfaces only	Subject can ambulate independently on level surfaces but requires supervision or physical assistance to negotiate any of the following: stairs, inclines, or non-level surfaces.
5 – Ambulator Independent	Subject can ambulate independently on non-level and level surfaces, stairs, and inclines

ANNEX C - INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE - SHORT FORM (IPAQ-SF)

We are interested in finding out about the kinds of physical activities that people do as part of their everyday lives. The questions will ask you about the time you spent being physically active in the **last 7 days**. Please answer each question even if you do not consider yourself to be an active person. Please think about the activities you do at work, as part of your house and yard work, to get from place to place, and in your spare time for recreation, exercise or sport.

Think about all the **vigorous** activities that you did in the **last 7 days**. **Vigorous** physical activities refer to activities that take hard physical effort and make you breathe much harder than normal. Think *only* about those physical activities that you did for at least 10 minutes at a time.

1. During the **last 7 days**, on how many days did you do **vigorous** physical activities like heavy lifting, digging, aerobics, or fast bicycling?

_____ days per week

_____ No vigorous physical activities → ***Skip to question 3***

2. How much time did you usually spend doing **vigorous** physical activities on one of those days?

_____ hours per day

_____ minutes per day

_____ Don't know/Not sure

Think about all the **moderate** activities that you did in the **last 7 days**. **Moderate** activities refer to activities that take moderate physical effort and make you breathe somewhat harder than normal. Think only about those physical activities that you did for at least 10 minutes at a time.

3. During the **last 7 days,** on how many days did you do **moderate** physical activities like carrying light loads, bicycling at a regular pace, or doubles tennis? Do not include walking.

_____ days per week

_____ No moderate physical activities → ***Skip to question 5***

4. How much time did you usually spend doing **moderate** physical activities on one of those days?

_____ **hours per day**

_____ **minutes per day**

_____ Don't know/Not sure

Think about the time you spent **walking** in the **last 7 days**. This includes at work and at home, walking to travel from place to place, and any other walking that you have done solely for recreation, sport, exercise, or leisure.

5. During the **last 7 days**, on how many days did you **walk** for at least 10 minutes at a time?

_____ **days per week**

_____ No walking → ***Skip to question 7***

6. How much time did you usually spend **walking** on one of those days?

_____ **hours per day**

_____ **minutes per day**

_____ Don't know/Not sure

The last question is about the time you spent **sitting** on weekdays during the **last 7 days**. Include time spent at work, at home, while doing course work and during leisure time. This may include time spent sitting at a desk, visiting friends, reading, or sitting or lying down to watch television.

7. During the **last 7 days**, how much time did you spend **sitting** on a **week day**?

_____ **hours per day**

_____ **minutes per day**

_____ Don't know/Not sure

This is the end of the questionnaire. Thank you for participating.

ANNEX D - MODIFIED ASHWORTH SCALE

Ashworth Scale is a clinical measure, which tests resistance to passive movement on a joint and scores the muscle spasticity in patients with neurological conditions. The modified version included the score +1 (BOHANNON; SMITH, 1987).

Score	Description
0	No increase in tone
1	Slight increase in muscle tone, manifested by a catch and release or minimal resistance at the end of the Range of Motion (ROM) when the affected part(s) is moved in flexion or extension
1+	Slight increase in muscle tone, manifested by a catch, followed by minimal resistance throughout the remainder (less than half) of the ROM
2	More marked increase in muscle tone through most of the ROM, but affected part(s) easily moved
3	Considerable increase in muscle tone, passive movement difficult
4	Affected part(s) rigid in flexion or extension

ANNEX E - SYSTEM USABILITY SCALE (SUS)

The System Usability Scale (SUS) method was developed by (BROOKE, 1996) and is defined as a simple ten-item scale that provides an overview of the subjective usability assessment (BROOKE, 2013). The item is composed of 10 statements with the five variable options ranging from "strongly disagree" to "strongly agree", where only one option should be ticked in each question. When necessary, one of the researchers explained the statement to the volunteer so that there is no misunderstanding in the answers.

The SUS score (BROOKE, 2013) is made as follows:

- For each of the 10 items is a value ranging from 0 to 4;
- For odd items (which are positively formulated items) one should subtract the '1' response from the volunteer (response – 1);
- For even-numbered items (which are negatively formulated items) one should carry out the subtraction of '5' minus the given answer (5 - answer);
- All 10 scores of each user are summed and the value is then multiplied by 2.5 to obtain the overall SUS value, which can be vary from 0 to 100.

To improve understanding of the value of SUS, it can be converted into percentage through a process called normalization. The average 50% of SUS is 68, or 50% of people who evaluated the system considered it usable.

The questionnaire will be answered based on the gait assisted by the robotic walker.

	Strongly Disagree 1	**Disagree 2**	**Neutral 3**	**Agree 4**	**Strongly Agree 5**
1 - I think that I would like to use this system frequently.					
2 - I found the system unnecessarily complex.					
3 - I thought the system was easy to use.					
4 - I think that I would need the support of a technical person to be able to use this system.					
5 - I found the various functions in this system were well integrated.					
6 - I thought there was too much inconsistency in this system.					
7 - I would imagine that most people would learn to use this system very quickly.					
8 - I found the system very cumbersome to use.					
9 - I felt very confident using the system.					
10 -I needed to learn a lot of things before I could get going with this system.					