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High-density surface EMG to investigate muscle activity during standing: implications for the training of postural control with EMG biofeedback in the elderly

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Declaration

I hereby declare that, the contents and organization of this dissertation constitute my own original work and does not compromise in any way the rights of third parties, including those relating to the security of personal data.

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Abstract

By recording surface EMG using standard bipolar EMG, previous studies have demonstrated that elderlies tend to activate their postural muscles during standing with a higher degree of activity and for a prolonged duration compared with young adults. The EMG biofeedback technique has been widely used to reduce the excessive level of muscle activity in different fields, e.g. the prevention and reduction of low back pain. In this view, EMG biofeedback could be a potential tool to assist aged subjects in reducing the excessive muscle activity during standing balance. However, whether the greater, prolonged activation observed locally in the muscles of aged subjects reflects the activation of the entire muscle is still an open question. It is possible that differences in the activation of postural muscles with aging are more or less expressive than previously appreciated. This thesis aimed at obtaining new insights into the rationale and the effects of the use of EMG biofeedback for the improvement of muscle efficiency during standing in the elderly. It was evaluated whether muscle activation during standing differ between young and aged subjects through a sophisticated detection system for the acquisition of surface EMGs from multiple regions of a single muscle (i.e., high-density surface EMG). Before to test this hypothesis, a methodological issue was addressed to verify whether high-density surface EMG is selective enough to detect during standing: (a) different activation between ankle muscles, as observed with other techniques (intramuscular electromyography); (b) variations in the activity within ankle muscles (i.e., soleus muscles). The results of this methodological study revealed that the medial portion of soleus muscle was activated continuously compared to the lateral portion of soleus and medial gastrocnemius, which were activated intermittently. These results suggest high-density surface EMGs can be used to discriminate the activity between ankle muscles (i.e., medial gastrocnemius and soleus) and muscle activity sampled from different regions of a single muscle (i.e., soleus) can provide estimates more representative of muscle activity during standing. High-density surface EMG was therefore used to assess muscle activity between young and aged subjects during standing. Key results indicate that during

standing: (a) tibialis anterior and medial gastrocnemius muscles were active for a longer duration in aged than young subjects; (b) a greater proportion of medial gastrocnemius volume was active in aged individuals. Collectively, these results corroborate previous evidence that elderlies tend to stand with a greater muscle effort than young subjects. Thus, the well-documented attenuation effect of EMG biofeedback on muscle activity may extend to the control of human standing posture with aging. This Thesis addressed additional issues which could be relevant to provide more representative EMGs of muscle activity to the subject through EMG biofeedback and to prove the attenuation effect of EMG biofeedback on the activity of lower limb muscles during standing. The following two questions were addressed: i) should EMGs be sampled from both lower limbs to provide more representative information about calf muscles activity? It was observed differences in muscle activity between left and right ankle muscle while young subjects stood at ease. These results indicate muscle activity should be sampled from the ankle muscles of both legs to avoid a biased recording and feedback of muscle activity during standing. ii) Is the attenuation effect of EMG biofeedback on the ankle muscles activity generalized to – or compensated by – other muscles during standing? These findings revealed the attenuation effect of EMG-audio feedback on ankle muscles is not compensated by other lower limbs muscles not included for the feedback. Therefore, the EMG biofeedback may be a promising technique to assist individuals in more efficiently controlling lower limbs muscles during standing. If the short-term, attenuation, effect of EMG-audio feedback on ankle muscles’ activity in young individuals observed here is generalized to other populations (e.g., the elderly) and retained after training, then, improvement in postural muscle efficiency may contribute significantly to an ability to maintain standing balance, to respond to unexpected perturbation, standing on narrow stances and walking; with potential implication for the prevention of falls.

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Chapter 1

Human Standing Balance, EMG and Aging

1.1 Human standing balance

Human beings usually stand upright to perform most motor tasks. Standing is essential for the majority of voluntary motor skills, such as walking, running, jumping, dancing and playing. Imagine dancers performing ballet movements, they often have to stand, stand on tip toes, stand on one leg or jump. Now, imagine a person doing daily activities. Probably this person performed different motor tasks while standing throughout the day, such as cooking or cleaning the house. People succeed in performing most functional tasks thanks to our ability to assume different body positions in space (i.e., postures) establishing a vertical orientation, commonly referred to as *postural control*. The possibility of performing a repertoire of postures during standing demands body stability.

In the human standing posture, body stability or balance is the ability of neuromuscular system to control the vertical projection of the body center of mass within the limits of a small base of support (i.e., feet; [1]). Once the body center of mass sways out of the limits of support base, the individual inevitably loses its balance. *Could the vertical projection of body center of mass move away from the limits of support base?* When people stand upright, the gravity pulls the body forward. It occurs since our body center of mass is often projected in front of the ankle joint due to the arrangement of body segments. Such an anterior sway of body must be compensated in the opposite direction (i.e., backward) to maintain body stability, otherwise the vertical projection of the body center of mass would fall off the limits of support base and individual would lose its balance. Then, the standing posture is characterized by some sort of movement, i.e. spontaneous, continuous and small body sways [2,3].

Efforts have been made to understand how humans control their posture during standing. There is a consensus in the literature that the control of standing posture relies on passive and active contribution of skeletal muscles. In contrast to some evidences [4], studies have demonstrated the intrinsic muscle stiffness contributes partially (from about 67% to 90% of the load stiffness; the gravitational toppling torque) to stabilize body sways [5,6]. Thus, the active contribution of skeletal muscles seems of marked relevance to maintain body stability during standing. Indeed, muscle activity have long been observed while subjects stand upright [7]. Insights have been gained into the active control of human standing balance from a technique referred to as *electromyography*. The following sections of this chapter provide brief introduction and recent methodological issues regarding the electromyography and its contribution to the recent knowledge of the control of standing balance even with aging.

1.2 EMG and the control of standing balance

The EMG signal. Electromyography is a technique for the detection and analysis of electrical potentials produced during muscle contractions (i.e., the electromyogram or EMG signal). The electrical potentials observed from the EMG signal reflect the functional units' activity of the neuromuscular system. Concisely, the functional unit of the neuromuscular system is the *motor unit*. The motor unit consists in a single motor neuron, including its dendrites and axon, and the muscle fibers innervated by its axonal branches [8]. Once a motor neuron discharges an electrical potential, it goes toward the neuromuscular junctions and generates action potentials within the muscles fibers. In literature, the summation of muscle fibers' activity in the motor unit is referred to as motor unit action potential.

1.2.1 How electromyography has been used in the literature of human postural control?

Surface and intramuscular EMG. EMGs are usually collected from electrodes positioned on the skin region over the muscle (surface electromyography) or inserted in the muscle tissue (intramuscular electromyography). Both techniques provide different advantages. On one hand, intramuscular electromyography has high selectivity for individual motor unit action potential in relation to surface electromyography, since intramuscular electrodes (e.g., needles or wires) are inserted close to the muscle fibers [9]. The action potentials of single motor units therefore can be easily observed from intramuscular EMGs, at least at moderate force levels (cf. Figure 1 in [10]). Then, intramuscular electrodes are often used to measure motor unit activity. On the other hand, detection systems for the acquisition of surface EMGs are expected to provide a more representative view of muscle activity, since surface electrodes sample from a larger muscle volume than intramuscular electrodes. The surface EMG signal normally conveys many action potentials from a population of active motor units [10]. Due to its non-invasiveness, surface EMG rather than intramuscular EMG has been broadly used in the literature. Recently, a methodological issue in the acquisition of surface EMGs for the understanding of active control of posture consists in how to use different detection systems for the sampling of surface EMG from postural muscles.

Recent findings from high-density surface EMG to bipolar surface EMG. A single pair of surface electrodes has been typically used for the sampling of EMGs from individual muscles [11–15]. Briefly, this acquisition procedure consists in: (1) determining and amplifying the voltage (potential difference) between the signal detected from surface electrodes and the reference electrode by a device, known as differential amplifier; (2) filtering such differential signal to attenuate possible undesired frequency components (e.g., 50 or 60 Hz) and converted into digital voltages values to be stored on a computer; (3) finally, surface EMGs are processed for estimating electromyographic indices (Figure 1; [10]). Due to the task of acquiring surface EMG data from the individual is unobtrusive and less labor intensive with a small number of non-invasive electrodes, this type of EMG recording referred to as bipolar surface EMG, has attracted several researchers and clinicians.

Bipolar surface EMG is often collected with a pair of electrodes closely spaced (i.e., inter-electrode distances ranged from 2 to 3 cm), which is commonly positioned approximately on the most prominent bulge of the muscle belly [11–15]. Such procedure is likely adopted in order to avoid factors that strongly influence the amplitude of surface EMGs, such as the presence of innervation zones and/or muscle tendons, and crosstalk; the recording of muscle activity other than that of interest [16]. Once collected in this way, it has been assumed that bipolar surface EMG represents the activity of entire muscle. However, recent literature has pointed out misunderstandings when using a single pair of shortly spaced electrodes, for example, to estimate the activity of muscles which play a predominant postural role.

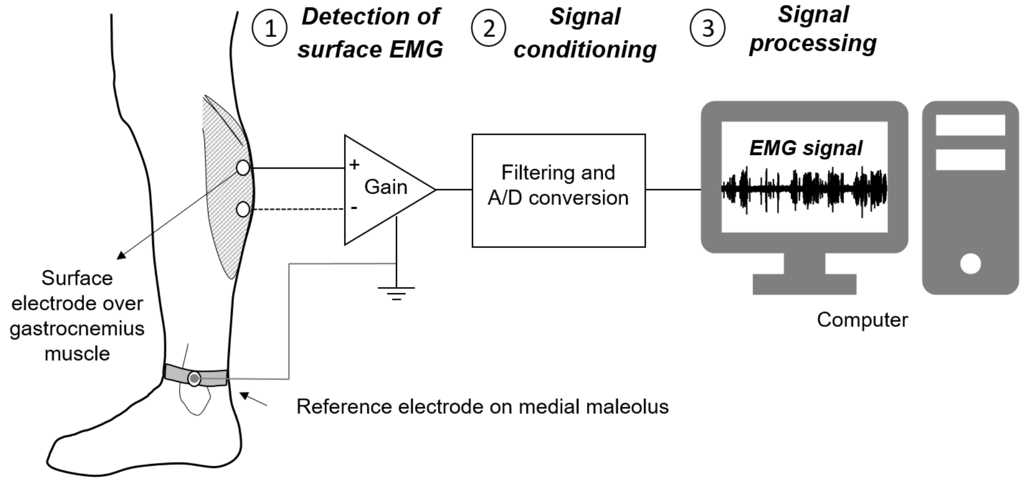

Figure 1: Simplified block diagram for the acquisition of bipolar surface EMG. The main steps for the acquisition of bipolar EMGs are: (1) detection and amplification of differential, bipolar surface EMG signals; (2) filtering and converting into digital voltage values; (3) signal processing to compute electromyographic indices or features.

The majority of muscles which have a postural role are large and pennate in-depth direction. For example, the fascicles of gastrocnemius and biceps femoris muscles are extended from the superficial to deep aponeurosis [17–19]. This muscle architecture implies the muscle fibers are not parallel to the skin and, consequently, action potentials do not propagate along the muscle. Action potential propagates toward the superficial aponeurosis and toward the deep aponeurosis, starting from the position of innervation zone. Then, a single pair of shortly spaced electrodes positioned over pennate muscles likely records surface EMG from a local muscle region. High representativeness of pennate muscles' activity from a couple of electrodes closely spaced depends on previously knowledge whether a localized muscle region (e.g., muscle belly) is activated during a given a task, e.g. standing.

Different regional patterns of activity within postural muscles (e.g., gastrocnemius), however, have been observed while subjects stand at ease. Such an uneven distribution of activity in the muscle have been usually appreciated from sophisticated detection systems based on multiple surface electrodes. *High-density surface EMG* is the general terminology used to indicate the sampling of multiple surface EMGs from a single muscle [10]. By collecting surface EMGs from different portions along the longitudinal axis of medial gastrocnemius with a linear array of surface electrodes (Figure 2A), for example, it is possible to observe the spatial variation in medial gastrocnemius' activity while a young subject stood at ease. More specifically, raw surface EMGs with relatively high amplitude were detected in the distal-medial region of medial gastrocnemius (channels 4-15) while EMGs with relatively low amplitude were sampled by channels 1-3 (Figure 2B). Close inspection of figure 2C shows that different action potentials were detected from different channels in the array. Once fascicles are orientated more parallel to the skin closer to the junction between medial gastrocnemius and Achilles tendon, propagating potentials may be observed in the distal region of this muscle. In such case, most distal electrodes (i.e., channels 13-15) sample from the same group of fibers rather than from different proximo-distal fibers [20]. Actions potentials detected by other channels in the array, thus, likely belong to different motor units. By comparing multiple intramuscular and surface EMGs sampled from different regions of gastrocnemius muscle, Vieira et al. [17] have demonstrated that the active motor units within gastrocnemius while subjects stand at ease likely occupy small territories along the muscle's longitudinal axis (~4cm), suggesting that surface EMG recordings are selective to reveal spatial variations in gastrocnemius' activity. Moreover, these authors detected a greater amount of active motor units in the distal-medial muscle portion compared with the proximal region. Likewise, by collecting surface EMGs from multiple gastrocnemius' regions with an array of surface electrodes, Hodson-Tole et al. [20] observed that gastrocnemius' postural motor units tend to be localized more distally (i.e., ~above the most distal part of gastrocnemii junction). On one hand, based on these evidences, one could argue that a couple of electrodes, not closely spaced, could provide bipolar EMG representative of medial gastrocnemius' activity during standing if positioned on the distal-medial regions of this muscle. On the other hand, a couple of electrodes shortly spaced would not suffice for the sampling of representative EMG activity from postural muscles, e.g. gastrocnemius. High-density surface EMGs seems to provide therefore more representative view of muscle activity in pennate muscles.

Are there alternatives for the acquisition of representative bipolar EMGs? Of more recent and applied interest is the notion of possible alternatives for the acquisition of bipolar EMGs which provide more representative estimates of postural muscles' activation. For example, the increase in the distance between the couple of electrodes. By combining surface and intramuscular electromyography, from a preliminary study, Vieira et al. [21] observed inter-electrode distances up to 5-7 cm may provide bipolar EMGs representative of medial gastrocnemius' activity with negligible soleus crosstalk. Then, increasing the distance between a couple of electrodes and positioning them in the distal-medial region of medial gastrocnemius (i.e., where likely reside most of active postural units, see Figure 2), might be an alternative to high-density surface EMG for the sampling of EMG signal more representative of medial gastrocnemius' activity while young subjects stand at ease.

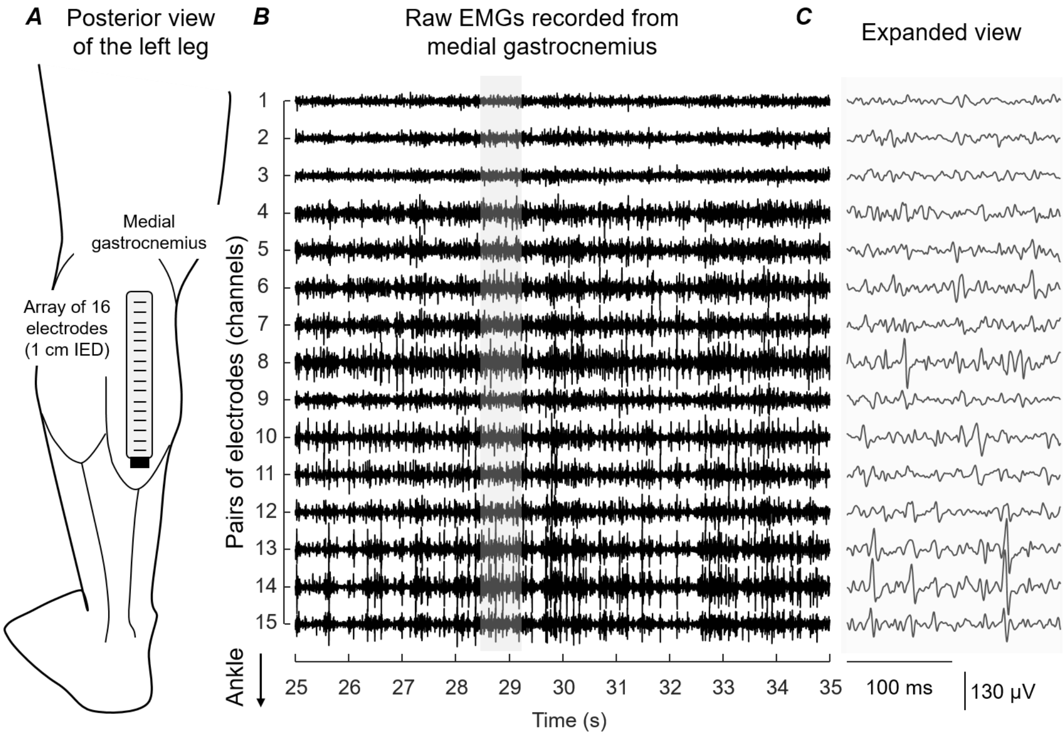

Figure 2: Spatial variation in medial gastrocnemius' activity during standing. (*A*) A linear array of surface electrodes (1 cm inter-electrode distance; IED) positioned on the skin region covering the medial gastrocnemius muscle of the left leg while a young subject stood at ease. (*B*) Raw surface EMGs collected in single-differential derivation from the medial gastrocnemius during 10s of standing. Note EMGs with relatively high amplitude were detected by 11 channels (channel 4 to 8 and channel 10 to 15). Relatively low modulations in EMG activity were recorded by channels 1-3 and 9 (i.e., four channels). (*C*) An expanded view (i.e., 500 ms) of all EMGs. Note signal propagation in channels 13-15.

Another possibility for the sampling of more representative bipolar EMGs could be to position two pairs of surface electrodes over the medial-distal region of medial gastrocnemius. There also are alternatives for the acquisition of more representative bipolar EMGs from other postural muscle, such as soleus. Previous studies have showed medio-lateral differences in soleus activation in different contexts [22,23]. Thus, a pair of surface electrodes positioned on each medio-lateral portion of soleus (i.e., the soleus portions which are not covered by gastrocnemius heads) could be relevant to detect a more representative view of soleus activity in case of possible spatial differences in activity within soleus.

These alternatives for the acquisition of more representative bipolar surface EMGs from ankle muscles with a parsimonious system of electrodes likely have potential implications in rehabilitation and training protocols. Systems for the acquisition of bipolar EMGs has been widely used, for example, as the core of a technique known as EMG biofeedback which is used to re-educate muscle activation [24,25]. Such technique has been recently proposed for the training of postural control (see Chapter 2). The knowledge and the possibility of sampling bipolar EMGs reflecting as much as possible the activation in the muscle as a whole seem to be crucial to avoid a biased view of muscle activity. Even though current issues from high-density surface EMG to bipolar surface EMG presented here, both techniques have provided insights into the understanding of human postural control.

1.2.2 Insights into the optimal, active control of standing posture from electromyography

Applications involving electromyography in the literature of human postural control generally consist in investigating: (i) which and how muscles are activated during standing; (ii) the mechanisms underpinning the control of standing posture; (iii) the effect of age and dementias on the postural muscles' activity during standing; (iv) the effect of training programs on postural activity. The priceless information obtained from these applications have drawn attention to a paradigm of optimal, active, control of standing balance; body stabilization with minimal muscle effort. The following lines summarize the studies that have indicated potential sources leading to such paradigm.

For more than fifty years, studies have demonstrated that specific muscles seem to contribute actively for the maintenance of standing balance. By sampling bipolar EMGs from different lower limbs muscles while young subjects stood at ease, Joseph et al. [7] observed clearly action potentials (amplitude ranged from 50 to 100 μV_{pp}) for the gastrocnemius and soleus muscles while for the thigh muscles (i.e., quadriceps and hamstrings), modulations in the amplitude of surface EMGs were usually close to the noise level (2 μV_{pp}). Even though Joseph et al. [7] have reported marginal contribution of thigh muscles for standing balance, recent studies have detected somewhat high degree of thigh muscles' activity while young subjects stood at ease [14] and observed that such degree seems to increase progressively with aging [11]. Extending the previous results, Heroux et al. [26] observed the calf muscles contributed differently while subjects stood at ease, in particular gastrocnemius' heads. By collecting intramuscular EMGs from gastrocnemius' heads (medial and lateral) and soleus muscle of seven subjects, these authors rarely observed motor units' activity in lateral gastrocnemius, while motor units' activity was identified within the medial gastrocnemius and soleus muscles. A forward lean on the verge of eliciting a step response was required to elicit moto units' activity in lateral gastrocnemius. Collectively, these findings demonstrate that the active contribution of calf muscles is crucial for the control of standing balance.

Besides evidences on a substantial degree of calf muscles' activity, studies have observed calf muscles are recruited intermittently in the standing posture. By collecting intramuscular EMGs from different regions along the longitudinal axis of medial gastrocnemius, Vieira et al. [27] showed that the active contribution of medial gastrocnemius to the ankle torque in standing was preferentially due to the intermittent recruitment of motor units. The intermittent motor units' recruitment occurred with a modal interval of ~500 ms, i.e. two recruitments per second. These two adjustments per second are consistent with tiny alterations in gastrocnemius and soleus movements for the compensation of the bodily forward sways. Loram et al. [28] observed that gastrocnemius and soleus active movements are orthodox, impulsive and sluggish (~400 ms) during standing balance. Regardless of potential sources account for such intermittent calf muscles' activation, e.g. internal planning [28] or event-triggered [29], timely postural activation reflects an efficient, active control of standing balance by the central nervous system.

The insights that normal population activates efficiently, not continually and intensely, their postural muscles during standing is congruent with the recent notion that the control of posture is primarily concerned with minimizing muscle activation. Recent studies have drawn attention to alternating periods of muscle activation and silencing (i.e., intermittent open loop control) seem to be a natural strategy of choice adopted by the central nervous system due to potential benefits: (1) allow the nervous system to sense joint angles without interference from muscle contraction [30]; (2) compensate for delays in the feedback loop [31]; (3) reduce metabolic costs [29]; (4) not increase postural instability [31]. As recently put forward in the literature, postural control system seems to prioritize the minimization of muscle activation at the cost of increased, though not excessive, postural sways [32]. In the view of optimal postural controller based on minimal muscle effort, learning to efficiently activate the postural muscles could be advantageous for the control of posture in individuals who stand with excessive muscle activation, e.g. elderly subjects.

1.3 Age-related changes in the active control of standing balance

Of recent interest is the notion of age-related alterations in postural muscles' activity during standing. From bipolar surface EMGs collected from the plantar flexors, previous studies have reported the elderlies tend to stand with a more continuous and higher degree of activation than young individuals. Moreover, while tibialis anterior is typically silent in young subjects during standing [33,34], in aged individuals it is often recruited [14]. Then, changes in the level of agonist/antagonist ratio have been often observed with aging. Benjuya et al. [11], for example, collecting bipolar surface EMGs from soleus and tibialis anterior muscles, observed the soleus/tibialis anterior activity ratio was significantly lower for the aged subjects when compared with the young individuals during standing, suggesting increases in the level of muscle cocontraction at the ankle in the elderly. Besides the differences in the ankle muscles' activity, the contribution of thigh muscles for standing balance seems to change with aging. By collecting surface EMGs locally from thigh muscles, Laughton et al. [14] observed aged subjects activated for a longer duration the vastus lateralis than young individuals during standing. Alternatively, Benjuya et al. [11] detected a higher degree of semitendinosus' activity (hamstring) while elderly individuals stood at ease with respect to young subjects. Collectively, the higher degree of calf and thigh muscles' activity for a prolonged duration in aged than young subjects suggest standing becomes progressively more demanding with aging.

The excessive muscle effort may be detrimental for postural control in the old age. As discussed previously, some ankle muscles are rarely activated while young subjects stand at ease (e.g., tibialis anterior). Such muscle silencing, however, seems to indicate the tibialis anterior plays a proprioceptive role for the control of standing balance. By recording EMG signals and ultrasound images during standing, Di Giulio et al. [33] revealed that changes in muscle length of tibialis anterior was markedly associated with alterations in the ankle joint angle, suggesting tibialis anterior may have an important, passive, postural role; it may be a better source of proprioceptive information than active muscles in the standing posture (e.g., calf muscles). The insight that silent muscles may have a strong proprioceptive role during standing implies aged subjects, when activating intensely their dorsal and plantar flexors for a long duration, are predicted to sense postural sway with interference from muscle contraction [30]. Prior account, indeed, has reported lower efficacy of proprioceptive pathway in aged than young subjects for the control of standing balance [35]. Besides a less economical, fatigue-resistant control [36], excessive muscle activity during standing might compromise the sensitivity of proprioceptive feedback for the control of posture. Therefore, the pattern of muscle activation should be considered in the repertoire of protocols so far designed for the improvement of postural control to understanding whether aged subjects are able to activate more efficiently their postural muscles during standing.

1.4 Available protocols for the training of postural control with aging

Generally, declines in the control of posture with aging has long been indicated due to the high incidence of falls; a major health problem for the elderly. Falls in elderly people have resulted in serious, different types of fractures (e.g., hip fractures) and physical injuries. Then, fall-related hospitalization is particularly common in this population [37–39]. Besides to cause individual morbidity, researchers have also stressed that falls can lead to declines in independence of activities of daily life and placement in assisted-living facilities, having potential impacts in the life of people 65 years and older [40,41].

Age-related alterations in the neuromuscular and sensory systems have been thought to contribute to an increased likelihood for falls. Briefly, advancing age has been associated with: (i) reductions in ankle muscle strength [42]; (ii) slow muscle response when compensating for sway perturbations [43,44]; (iii) excessive ankle muscle coactivation for standing [11–13]; and (iv) declines in function within sensory systems [45]. Such changes in the postural control systems lead to increases in the size of postural sway while aged subjects stand upright [45,46]. Increases in the amplitude of body sways might compromise upright balance and can lead to falls [47,48]. Thus, several training programs have been proposed primarily to preserve and/or reduce postural sway within functional levels. These training programs are usually referred to as balance training. Some examples of balance training are shown as follows:

One or two legged exercises on unstable support surfaces. Conventional balance training focus on challenging body stability while individuals stand on unstable support surfaces with one or two legs [49]. Different training devices are used as unstable support surfaces, such as wobbling boards, tilt boards, spinning tops, half discs, soft mats and cushions (cf. Figure 1 in [50]). An alternative to these training devices is keeping the balance control on a tightened ribbon, sport activity known as "slacklining". Studies have shown balance training with unstable support surfaces leads to reductions in the amplitude of postural sway. For example, by applying the balance training based on single limb stance on an ankle disk in athletes who swayed excessively due to ankle joint injury, Gauffin et al. [51] observed that these subjects were able to reduce the size of their postural sway (~3 mm) while standing on the symptomatic foot, suggesting an improved body stability after this training. Keller et al. [52] observed individuals who attended slackline training could decrease their postural sways while standing on a free-swinging platform with one leg and maintain balance on the slackline for at least 20 s after training. Elderly subjects were also able to reduce the amount of postural sway following a balance training program [53].

Besides reductions in the postural sway size, balance training exposing individual to unstable support surface may: (i) increase the muscle strength [53]; (ii) improve jumping abilities [54]; (iii) increase maximal rate of force development [55]; and (iv) decrease the level of ankle muscles' activity during standing [53,56]. Then, this type of training is also frequently used in prevention and rehabilitation of the lower limb muscles, e.g. in athletes. More recently, studies have indicated this type of balance training induces plasticity of the sensory-motor pathways [49,53], contributing likely for a more efficient control of posture by means of reduction in the neural drive to the muscle.

Human bodily practices. The effects of human bodily practices as martial arts (e.g., *Tai Chi* and *Qigong*) and *Yoga* on aged postural control has been recently investigated in literature, since they demand the performance of several postures with relatively slow, meditative, dance-like movements during standing (for a review see [57,58]). Prior accounts have reported aged subjects who practice these types of "standing meditation postures" may improve their balance skills and, consequently, they are less likely to falls [59–61]. For example, Li et al. [60] observed the Tai Chi training was more promising to reduce fall risk with respect to stretching training. More specifically, these authors reported a small number of falls, injurious falls and reduced fear of falling after 6-month in elderly subjects who attended Tai Chi training than those following stretching training. In addition, Tai Chi group obtained greater score in Berg Balance Scale than stretching group; this result suggests Tai Chi contributed for improvements of balance skills in different functional tasks. Additionally, a recent study observed Tai Chi training may be effective to reduce the size of postural sway during standing (Sheng et al., 2014).

On the view of decrease risk for falling and improvements of balance skills, human bodily practices (e.g., *Tai Chi, Qigong* and *Yoga*) have been indicated as promising training of postural control [62].

Biofeedback of postural sways. Motor tasks based on the precise control of body position in space from the feedback of variables related to human motion have been proposed recently as a promising component in balance training programs. More specifically, biofeedback techniques for balance training involve typically the acquisition of variables related to balance performance (e.g., joint angles, trunk acceleration or the position of body center of pressure) and feed back to the subject, allowing thus individuals to have an augmented information about their own body sways and control them voluntarily [63,64]. Among available biofeedback-based balance training, information can be displayed back to the subject from different modalities, such as audio or visual [63,65]. The protocol which is widely used in rehabilitation programs is the visual feedback of center of pressure (CoP) position while individuals stand at ease [66]. From the visualization of CoP signal, subjects are expected to reduce their postural sways during standing [67]. More specifically, force plates are often used for the quantification of CoP position during standing (i.e., the resultant vector of ground reaction forces under the feet; [47]). With aging, individuals who attended training programs related to CoP visual biofeedback could learn to maintain upright stability with smaller body sways [66,68]. Reductions in the size of postural sway were also achieved by the elderly from wearable systems for body motion-based audio biofeedback [69,70]. Most importantly, although not very clear, improvements in balance performance after the biofeedback-based training protocols likely reflect in a reduced incidence of falls. Significant effect of the visual feedback-based balance training compared with control group was observed on recurrent falls (8% vs 55% of falls) during a 1-year follow up period as well as a reduced risk of falling [71].

Recently, an alternative biofeedback approach based on biological-related signals, i.e., surface EMGs, has been proposed for balance training [72]. On the view of optimal postural control based on minimal muscle effort and the potential impact of EMG biofeedback technique in reducing the excessive muscle activity, EMG biofeedback-based balance training is a promising approach for the improvement of postural control and, consequently, prevention of falls in the elderly. The following chapter provides brief overview about the significant effects of EMG biofeedback on the pattern of muscle activity of aged and young subjects in different contexts, besides its potential impact in reducing the unnecessary muscle effort while individuals stand at ease.

Chapter 2

EMG biofeedback for the training of postural control

2.1 EMG biofeedback to re-educate muscle activation

Biofeedback technique has been applied since at least the 60s, to train and/or allow subjects to control biomechanical or biological variables which are not under direct control [66,73]. On one hand, biomechanical biofeedback is related to variables of body motion (e.g., joint angle or CoP position), which are commonly used for balance training, as discussed previously (see chapter 1, subsection 1.4). On the other hand, biological-related variables include, for example, the electrical activity of muscles (e.g., bipolar surface EMG), brain and heart [73]. Several researchers have reported the EMG biofeedback is a promising technique to suppress the excessive level of agonist activity or muscle coactivation [74–76], with potential implications for the improvement of postural control in terms of muscle efficient, e.g. with aging.

One of pioneers in suggesting biofeedback as potential tool for the training of muscle activity was Basmajian et al. [77], testing a biofeedback protocol for the training of single motor units within a hand muscle. By sampling intramuscular EMGs from a hand muscle (i.e., abductor pollicis brevis) and feed back them to the subject in the form of auditory or visual signals, the author reported volunteers were able to learn how to: (i) repress the activity of a single motor unit and recruit another one; (ii) recruit the several units over which they have gained control; (iii) reduce and increase the discharge frequency of a well-controlled unit. After biofeedback training, most participants learnt to control the activity of individual motor units even in the absence of feedback. From these evidences, several researchers have widely accepted the idea of using biofeedback technique to assist subjects with impaired motor control in recovering the normal pattern of muscle activity [78]. Bipolar surface EMG rather than intramuscular EMG, however, are often used within the designs of EMG biofeedback techniques due to its non-invasiveness and simplest form to detect surface EMGs.

Briefly, EMG biofeedback design involves typically three technical stages to transmit feedback based on surface EMGs to the subject: (1) the sampling of EMG signal from a couple of surface electrodes; (2) the estimation of the level or timing of muscle activity (i.e., descriptive signal of raw surface EMG); (3) the representation of this signal to the subject through audio, visual or vibration cues (Figure 3). From table 1, for example, it is possible to observe that studies have commonly monitored the EMG level and feed back it to the subject from audio and/or visual modality. More recent biofeedback techniques, as shown in Figure 3, use wireless EMG devices to transmit bipolar surface EMGs to receivers such as notebook, tablet or smartphone. Within the receiver, the targeted information is processed and transmitted to the subject as biofeedback. These sophisticated and wearable EMG devices are thought to have a relevant potential for rehabilitation or training programs. Once these recent EMG systems are based on surface EMG recording and easy to use software to acquire and provide biofeedback information to the subject, biofeedback protocols can be easily implemented in the community setting, and potentially in patients' homes, contributing for a more frequent training. In addition, the task of providing EMG biofeedback information to subjects is less time consuming and less intrusive.

When using biofeedback techniques to continuously detect and provide real-time information about muscle activity to the subject, the main goals are (i) attenuation of unnecessary muscle activity and (ii) muscle activity facilitation [25,79,80]. From table 1, it is possible to appreciate the multiple pathologies the biofeedback techniques have been tested on. For instance, the EMG biofeedback has been effective to directly assist patients with chronic low back pain in reducing the excessive level of paraspinal activity during standing [75]. By providing the visual feedback of erector spinae' activity, collected from a pair of surface electrodes, to the patients within 15 treatment sessions spread over a period of three weeks, Nouwen [75] observed these patients were able to stand with relatively low levels of paraspinal activity after biofeedback training. Alternatively, reductions in the level of muscle activity at wrist joint were also observed in stroke patients following an EMG biofeedback training. Aslan et al. [81] applied 15 sessions of biofeedback training, lasting 20 minutes each one, aimed at reducing wrist flexor muscle spasticity at the hemiplegic arm in patients following stroke. The authors observed lower level of wrist flexors' activity of about 20% after training with respect to baseline measurements.

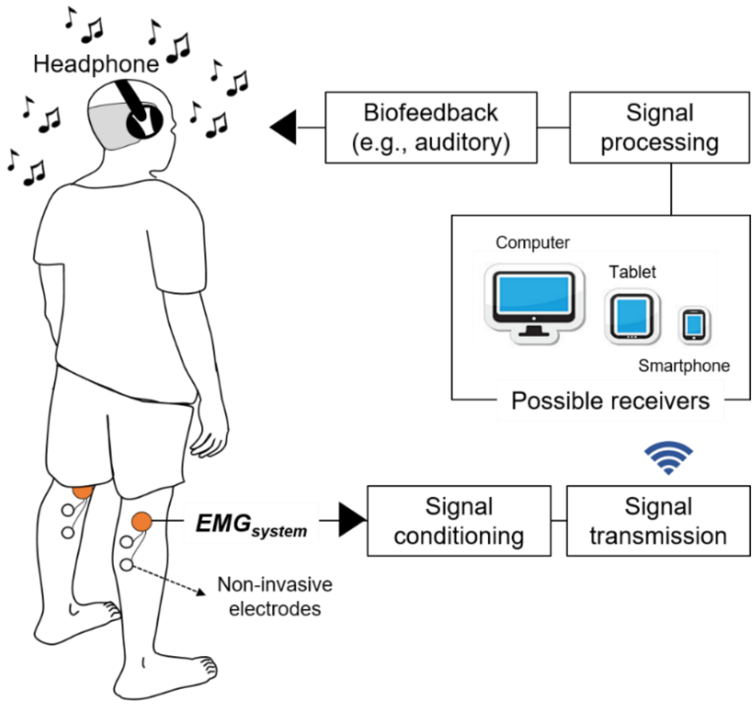

Figure 3: General block diagram of an EMG biofeedback system. EMGs from gastrocnemius muscles are sampled with surface electrodes from a wearable and wireless EMG system (orange circles on the body's subject in the panel). Such system transmits the EMG signals to computers, tablets or smartphones (receivers) which process and transform the EMG into an audio signal and provided it to the subject (EMG-audio biofeedback) through headphone.

Excessive muscle activity has also been attenuated when individuals are informed about the timing of muscle activity during a given task. By providing feedback when the level of muscle activity was higher than a threshold for a given duration, Hermens et al. [25] observed individuals with neck-shoulder myalgia were able to relax more frequently their trapezius muscles of about 15% after training with respect to pre-training during working conditions. These results suggest patients with different clinical conditions could learn to attenuate their muscle activity through the EMG biofeedback-based training. EMG biofeedback may also have the potential to facilitate the level of activity in individual muscles. For example, Ng et al. [80] compared the EMG ratio of vastus medialis/vastus lateralis between an exercise group and an EMG biofeedback+exercise group. The biofeedback protocol proposed by Ng et al. [80] aimed at assisting subjects to perform different exercises while increasing the level of activity in vastus medialis and maintaining a constant level of vastus lateralis' activity. These authors reported increases in the vastus medialis/vastus lateralis EMG ratio of about 30% after training. Collectively, these results indicate EMG biofeedback as potential tool to re-educate muscle activity.

Possible mechanisms may underpin the effectiveness of EMG biofeedback in alternating muscle activity. Two hypotheses come from the experiments of Basmajian [77] who observed subjects can recall voluntarily the activity of different single motor units while inhibiting the activity of neighbors after EMG biofeedback training. Basmajian [82] speculated either new pathways or auxiliary feedback pathways can be recruited following the EMG biofeedback training. Favoring the latter hypothesis, Wolf [83] suggested that the primary role of EMG biofeedback might be to activate central synapses previously unused or underused in executing motor commands. These potential sources accounting for the effectiveness of EMG biofeedback are still considered in recent literature (Huang et al. 2006). Regardless of possible mechanisms underpinning the effect of EMG biofeedback on muscle activation, subjects may be able to change their pattern of muscle activity through this technique.

Notwithstanding reductions/increases in muscle activity from the EMG biofeedback, it is not clear whether such changes in the muscle activity are retained after training. A common question raised by researchers is "*Are any benefits maintained after biofeedback training ceases?*"[84]. It seems well-accepted in literature different biofeedback training programs may be effective to improve muscle activity temporary, regardless of clinical application. The minimal training volume showed in Table 1, for example, was that one of Holtermann et al. [24]. More specifically, participants involved in this study attended five sessions of unilateral biofeedback training on trapezius activity during normal computer work at their workplace was carried out. Holtermann et al. [24] observed this short EMG biofeedback-based training was sufficient to reduce the degree as well as the timing of trapezius activity in computers workers during normal work. However, the long-term effect of the EMG biofeedback training is not commonly evaluated, e.g. for reduction of pain intensity in fibromyalgia [85]. Among the studies presented in table 1, only the studies of Hermens et al. [25] and Dannecker et al. [86] performed follow-up measurement. For example, Hermens et al. [25] reported the attenuation effect of EMG biofeedback was retained even after 4 weeks of EMG biofeedback training. The results of Hermens et al. [25] reflect learning has taken place in addition to short-term improvements in performance, with potential implications for rehabilitation process [84]. Therefore, it seems relevant to consider the assessment of retention effect for determining the effectiveness of biofeedback protocols by means of learning in different clinical applications.

2.2 EMG biofeedback for reducing the excessive muscle effort during standing

EMG biofeedback has been recently proposed as alternative biofeedback technique for the improvement of postural control in standing posture. Vieira et al. [72] investigated for the first time the influence of EMG-audio feedback protocol on the ankle muscles' activity during standing. With this feedback approach and in relation to standing at ease, these authors observed young subjects were able to decrease by 5% and 10% the degree of medial gastrocnemius and soleus activity, respectively. Most importantly, such a reduced muscular effort did not lead to larger postural sways and greater antagonistic activation (i.e., higher tibialis anterior's activity). These results reveal young individuals could attenuate the level of ankle muscles' activity, reflecting likely more efficient postural activation with the EMG-audio feedback without hindering postural stability. Additional investigations related to the effects of EMG biofeedback on postural activity, however, might be relevant to prove the effectiveness of this technique in reducing the level of postural activity during standing.

Researchers have long been indicated the assessment of response generalization of EMG biofeedback should be considered, that is, the effect of EMG biofeedback on muscles other than that included for biofeedback (i.e., targeted muscle) [87]. For example, Poppen et al. [88] stated "the biofeedback trainer should systematically observe and record the postural changes that occur throughout the trainees' body, and not focus narrowly on the EMG signal from a particular muscle group". The main concern is whether the attenuation effect in an individual muscle from EMG biofeedback is manifested at – *or compensated by* - other muscles during a given task [89]. Prior accounts have demonstrated that an attenuation effect and/or compensation effect from EMG biofeedback protocols may occur between muscles depending on the motor task. More specifically, the following effects have been observed when the EMG biofeedback is provided from a single muscle: (i) the

Table 1: The effect of EMG biofeedback (BF) - based training on muscle activity

Article	Subjects	BF variable	BF mode	Training*	Clinical application	% change (pre-post training)
Estella Ma et al. [79]	Young healthy	EMG level	Visual	2x wk, 4 wks[#] Total: 8 sessions	Hyperfunctional voice disorder	EMG level: ↓ 10%.
Aslan et al. [81]	Elderly hemiplegic	EMG level	Audio and visual	5x wk, 3 wks, 20 min DS Total: 300 min	Stroke	EMG level: ↓ 20%.
Chao Ma et al. [90]	Young with neck-shoulder pain	EMG level	Audio	2-5 wk, 6 wks, 120 min DS Total: 1440-3600 min	Low back and neck pain	EMG level: ↓ 50%.
Holtermann et al. [24]	Young healthy	EMG level	Audio and visual	1 wk, 5 wks, 30 min DS Total: 150 min	Low back and neck pain	EMG level: ↓ 50 %; EMG timing: ↓ 10 %.
Hermens et al. [25]	Young with neck-shoulder pain	EMG timing	Vibration	5 wk, 4 wks, 480 min DS Total: 9600 min	Low back and neck pain	EMG level: ↓ 10 %; EMG timing: ↓ 15 %.
Nouwen et al. [75]	Young with low back pain	EMG level	Visual	5 wk, 3 wks, 15 min DS Total: 225 min	Low back and neck pain	EMG level: ↓ 40 %.
Barney et al. [91]	Young healthy	EMG level	Audio	5 wk, 3 wks, 30 min DS Total: 300 min	Knee conditions	EMG level: ↑ 10 %.
Ng et al. [80]	Young with patellofemoral pain	EMG level	Visual	7 wk, 8 wks, 30 min DS Total: 1680 min	Knee conditions	Agonist ratio: ↑ 30 %
Christanell et al. [92]	Young	EMG level	Audio and visual	4 - 2 wk, 6 wks[#] Total: 24 – 12 sessions	Knee conditions	EMG level: ↑ 25 %
Dannecker et al. [86]	Young with stress urinary	EMG level	Visual	7x wk, 4 wks, 20 min DS Total: 560 min	Urinary incontinence	EMG level: ↑ 50 %

* Protocol of biofeedback-based training only. wk – number of weekly sessions; wks – number of weeks; DS – duration of training session; [#] DS not declared

transfer of the attenuation effect to the contralateral muscle [24,93]; (ii) limitation to transfer the relaxation effect to not-synergistic muscles [93]; and (iii) compensation effect (i.e., increases in the level of activity) on other muscles not included in the feedback during strictly controlled biomechanics conditions (i.e., isometric shoulder abductions [94]). These results suggest is relevant to consider the assessment of effect generalization to prove the effectiveness of innovative biofeedback protocols, e.g. in reducing the level of muscle activity during standing.

If EMG biofeedback is able to assist young individuals in more efficiently controlling leg muscle activity, it could be a promising technique to recover muscle activity within optimal levels in different populations or patients during standing. For example, it is well-documented aged individuals activate their ankle and thigh muscles to a greater extent and for a prolonged duration while stand at ease compared with young subjects [11,12,14]. With respect to healthy aged subjects, parkinsonians show exacerbated ankle stiffness and muscle coactivation during standing balance [95]. In the case of individuals with excessive muscle activity learn to control more efficiently their postural muscles from an EMG biofeedback-based balance training, thus, improvement in the postural muscle efficiency could contribute significantly to an ability to maintain standing balance, to respond to unexpected perturbation, standing on narrow stances and walking; with potential implication for the prevention of falls.

Chapter 3

Does the global temporal activation differ in triceps surae during standing balance? [1]

3.1 Abstract

One of the most important muscular groups which contribute to maintain standing balance is triceps surae. However, it is unclear whether the muscles which constitute the triceps surae, medial gastrocnemius and soleus, have different temporal patterns of activation during upright stance. The work described in this chapter aimed at evaluating whether the temporal activation differ among the postural muscles of triceps surae in young subjects during standing. Nine male volunteers performed two tasks: standing quietly and with voluntary back and forward sways over their ankle. Electromyograms (EMGs) from soleus medial and lateral regions and from MG were sampled with linear arrays of surface electrodes. The percentage of muscle activation in time (i.e. temporal index) was computed for each muscle during upright standing. The results revealed that the medial portion

[1] Scientific paper published in "37th Annual International Conference of the IEEE Engineering in Medicine and Biology Society", 2015; Authors: Dos Anjos F, Fontanella F, Gazzoni M, Vieira TM.

of soleus muscle was activated continuously compared to the lateral portion of soleus and medial gastrocnemius, which were activated intermittently. Therefore, the global temporal activation differed among the postural muscles of triceps surae during standing balance.

3.2 Introduction

Triceps surae activation is crucial for the control of human standing posture. As the human center of gravity is projected anteriorly to the ankle joint, plantar flexors must compensate the gravitational toppling torque to ensure standing balance [1,96–98]. Curiously, these agonist muscles seem to present distinct pattern of activation; small portions of soleus (SOL) are continuously activated while those of medial gastrocnemius (MG) are intermittently activated during upright stance [27,99,100]. Conversely, lateral gastrocnemius provides a marginal, active contribution to standing maintenance [26,101]. However, as intramuscular electrodes and bipolar electrodes sample from a localized muscle region, it is unknown whether the temporal activation between the main postural controllers of ankle, MG and SOL, differs globally. In this context, the evaluation of different SOL and MG portions may be essential to distinguish whether theses muscles are differently controlled by the central nervous system (CNS) during standing balance. Action potentials generated by different populations of motor units in a large calf muscle volume may be sampled with an array of surface electrodes [17,20], allowing to investigate whether the temporal pattern of activation observed locally for the SOL and MG during upright stance represents a larger muscle volume. This study aimed at evaluating whether the temporal activation estimated from a large muscle volume (i.e., global) differs between the medial gastrocnemius and soleus muscles during standing balance.

3.3 Methods

3.3.1 Participants

Nine male volunteers (mean±SD: 26±3 years; 72.33±9.43 kg body mass; 1.77±0.06 m height) provided written informed consent before participating in the study. All procedures used in this study were in accordance with the Helsinki Declaration of 1975, as revised in 2000. These subjects did not report any balance impairment, neurological disorders, muscular injuries or the intake of medications that could affect their standing balance.

3.3.2 Experimental procedures

Participants were instructed to stand upright on a force plate with their eyes open, their arms alongside the body and their feet in parallel and comfortable position. Previous evidence has shown a significant effect of intention on the standing control mechanisms [102]. As we wish to investigate the muscle activation during natural, unconscious standing, subjects were engaged in active conversation to get distracted from the task of controlling their standing body. Furthermore, participants were also asked to sway as much as possible back and forward over their ankles, without lifting their heels and toes. These standing tasks lasted 60s and data acquisition was initiated and ceased upon the issuing of the auditory cues "ready, go" and "stop", respectively.

3.3.3 Instrumentation

Linear arrays of surface electrodes (10 mm inter-electrode distance) were placed on MG and SOL muscles of dominant leg after carefully cleaning the skin with abrasive paste [103]. The positioning of electrodes on each muscle was guided through ultrasound imaging (Telemed, Echo Blaster 128, 2 MHz to 10 MHz linear probe). Specifically, for the MG muscle, EMGs were sampled with an array of 16 electrodes, placed along the length of MG, aligned with MG-Achilles tendon junction and with its most proximal electrode positioned 2 cm distally from the popliteal fossa (Figure 4). For the SOL muscle, a surface array of 4 electrodes was placed on each muscle medial (MSOL) and lateral (LSOL) portions in order to investigate different temporal pattern of activation. The medial and lateral arrays were positioned at an angle of 45 degrees to the leg longitudinal axis and with their most proximal electrode located 3 cm distally to the junction between MG and LG and the Achilles tendon, respectively. Such array orientation may facilitate the recording of small action potentials while standing and it is characterized by a high signal-to-noise ratio [23].

EMGs were recorded in single-differential derivation with a multi-channel amplifier (10-750 Hz bandwidth; EMG-USB, OTBioelettronica, Turin, Italy). All signals were sampled at 2048 kHz using a 12-bit A/D converter (± 2.5 V input dynamic range). EMGs were amplified by 5000-10000 to maximise signal-to-noise ratio and then offline, band-pass filtered (cutoff frequencies: 15-350 Hz) with a second order Butterworth filter. COP displacement was computed from the vertical force signals supplied by the force plate (9286AA Kistler, Zurich, Switzerland) and digitised synchronously with the surface EMGs.

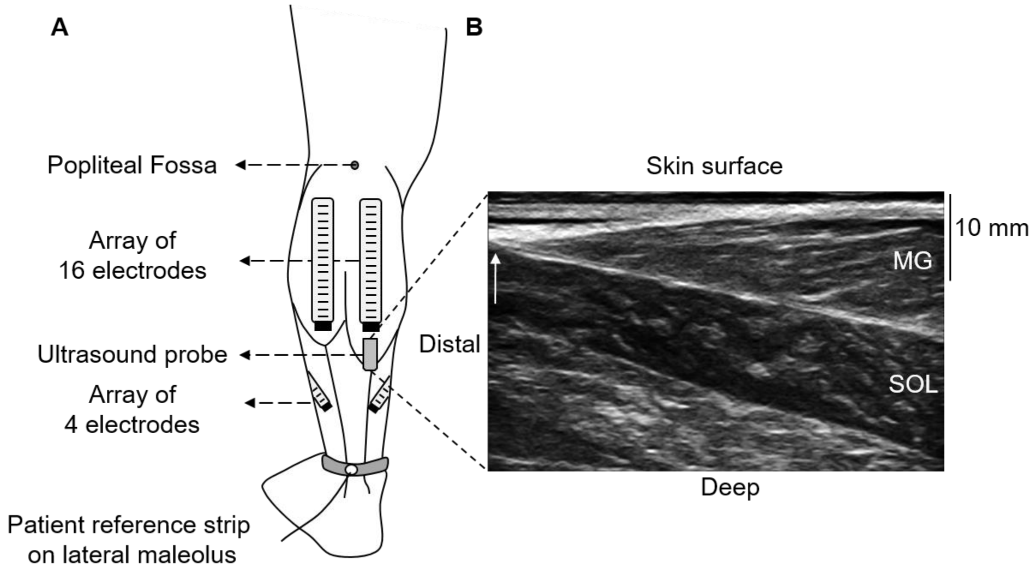

Figure 4: A: Schematic representation of the linear surface arrays positioned on MG and SOL while a subject is standing. B: MG-Achilles tendon junction (arrow point) identified in the ultrasound image to guide the placement of surface arrays.

3.3.4 Data analysis and statistics

The global temporal pattern of MG and SOL activation was quantified through a temporal index of muscle activation obtained from the root mean square (RMS) amplitude of EMGs.

RMS values were computed using a 40 ms running-window for each channel separately, providing 1500 RMS values per channel [14]. As calf muscle presents bursts of activity during upright stance, short epochs are more sensitive for the identification of ON or OFF states. For each RMS value, an ON or OFF state was assigned according to whether the RMS amplitude exceeded the baseline threshold for a given duration. More specifically, if at least five consecutive RMS values exceeded the baseline these five samples were deemed ON (i.e., RMS amplitude indicated an active state if they exceeded the baseline for at least 200 ms). This procedure was selected because it was observed to be not sensitive to spurious occurrences of an isolated, high RMS value and because previous evidence suggests bursts of calf muscles activity during standing occur in 200 ms intervals [27].

Following previous evidence (Laughton), baseline was defined as three standard deviations over the mean RMS amplitude and calculated over 3 s of muscle rest (75 epochs of 40 ms). On the view of deactivation of plantar flexors when the body sways closer to ankle joint (Di Giullio 2009 and Vieira 2012), periods of muscle rest were determined here during the voluntary sway task; calf muscles' EMGs were within the noise level during backward sways (Figure 5).

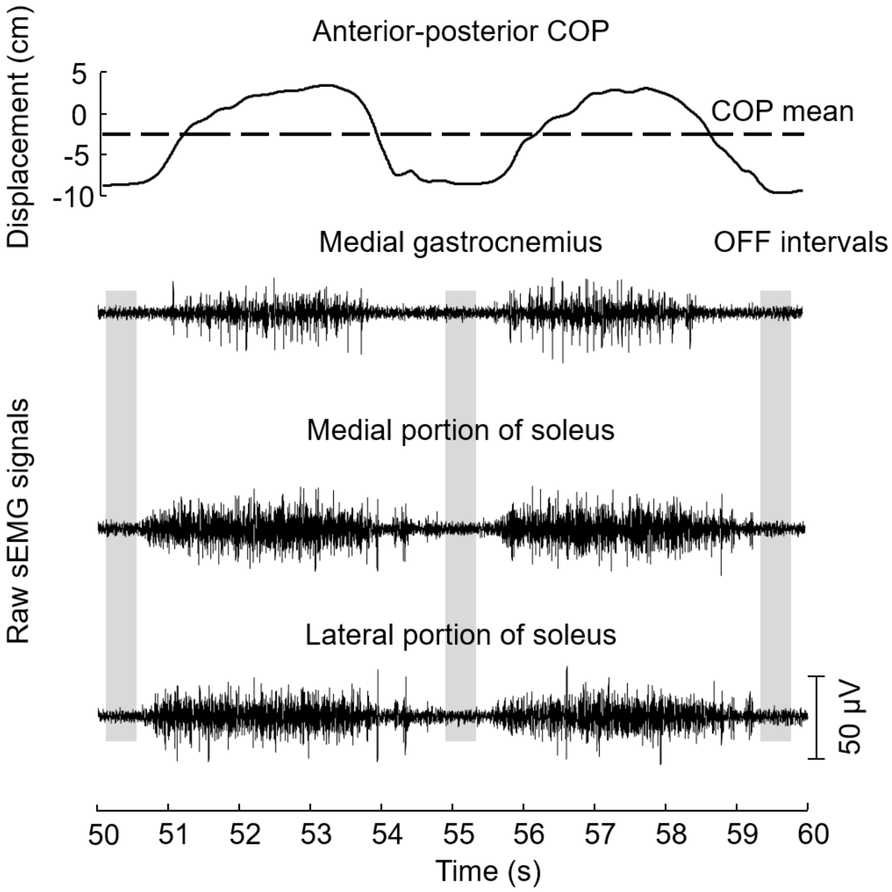

Figure 5: An example of fluctuations in calf muscles' EMG amplitude and in COP position during the voluntary sway task. EMGs in the intervals indicated with grey bar (backward sways) were considered to compute the baseline level.

As EMGs were recorded from an array of surface electrodes from each muscle, baseline was defined separately for each channel. ON-OFF levels of each channel were processed through the logical disjunction ("Or") to produce the resultant global on-off signal (temporal index). Consequently, whenever there was an ON level for any channel, the resulting state for the corresponding time sample within the 60s acquisition period was regarded as ON. Furthermore, as electrodes located in the distal portion of MG can detect the same propagating action potential, indicating the length of same group of muscle fibers instead of action potentials generated by motor units in different muscle regions [20], such channels were excluded from the analysis. The selection of the channels to exclude, showing only propagating action potentials, was performed by the experimenter through visual inspection of EMG signals. Figure 6 illustrates the ON-OFF calculation for EMGs collected from the MG muscle.

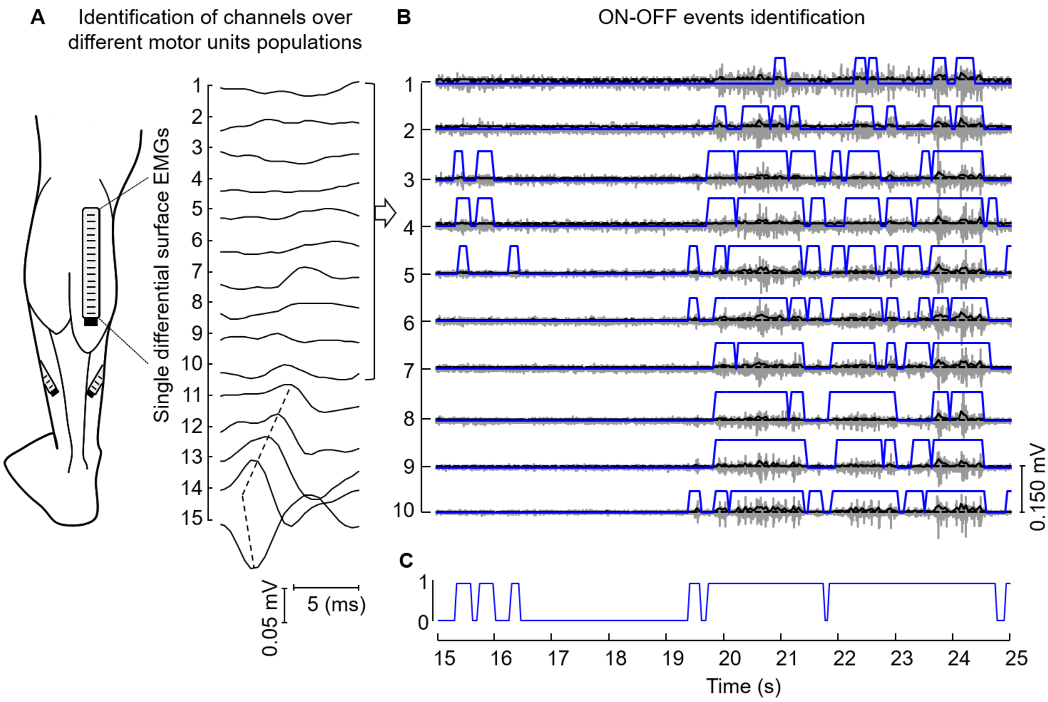

Figure 6: Calculation of the resultant ON-OFF signal of MG for a representative subject while standing. *A*: Identification of MG channels over different muscle fibers for MG ON-OFF calculation. *B*: ON-OFF computation for each channel selected from MG muscle; Raw EMGs (grey), RMS baseline (dashed line), RMS signal (black) and ON-OFF signal (blue). *C*: Resultant ON-OFF signal of MG calculated through logical disjunction ("Or") between EMGs from MG muscle.

The Kruskal-Wallis ANOVA by Ranks was used to verify whether the temporal index of muscle activation differed between muscles. Post- hoc comparisons of ranks for all groups were made considering the significance level of 5%.

3.4 Results

Distinct patterns of global activation in time were observed in triceps surae through the temporal index. The Kruskal-Wallis test revealed that the temporal indices were significantly different from each other ($p<0.01$). Post hoc comparisons further indicated a greater number of ON events for MSOL than for MG ($p<0.01$) and LSOL ($p<0.05$; Figures 7 and 8). Even those individuals who presented very low values of time index for the MSOL (46.19% and 50.60%; see Figure 8), obtained a smaller number of ON events for the MG (19.82% and 16.68%) and for the LSOL (18.35% and 26.43%), respectively. Moreover, there was no significant difference between the temporal index of MG and LSOL ($p>0.05$).

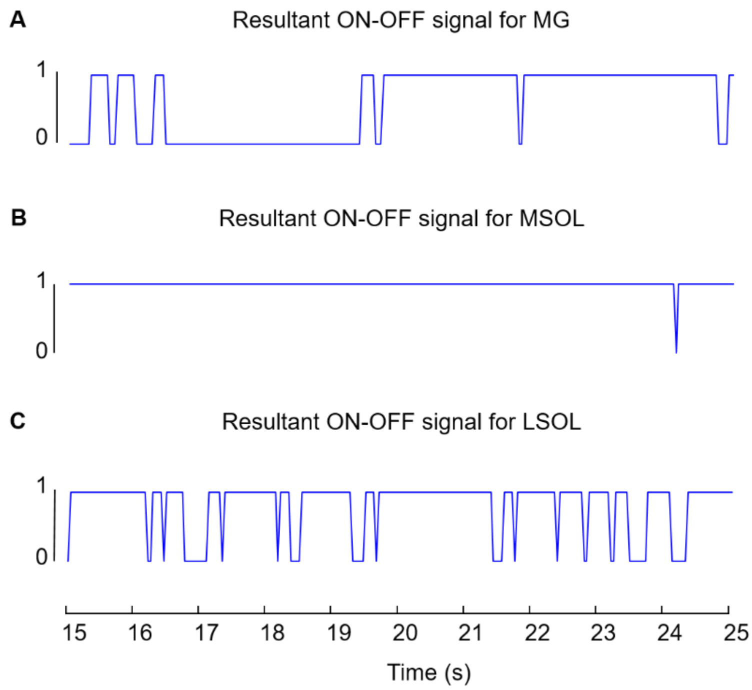

Figure 7: Resultant ON-OFF signal obtained for the MG (*A*), MSOL (*B*) and LSOL (*C*) muscles for a representative individual while standing.

3.5 Discussion

Prior studies have observed a heterogeneous pattern of activation in small portions of triceps surae, with SOL having a continuous activation and MG an intermittent activation [27,99,100,104]. Nevertheless, these results likely do not represent the temporal activation of these muscles globally, since they were evaluated from localized muscle regions. In this study the global temporal activation in the triceps surae of healthy young men was investigated during upright stance.

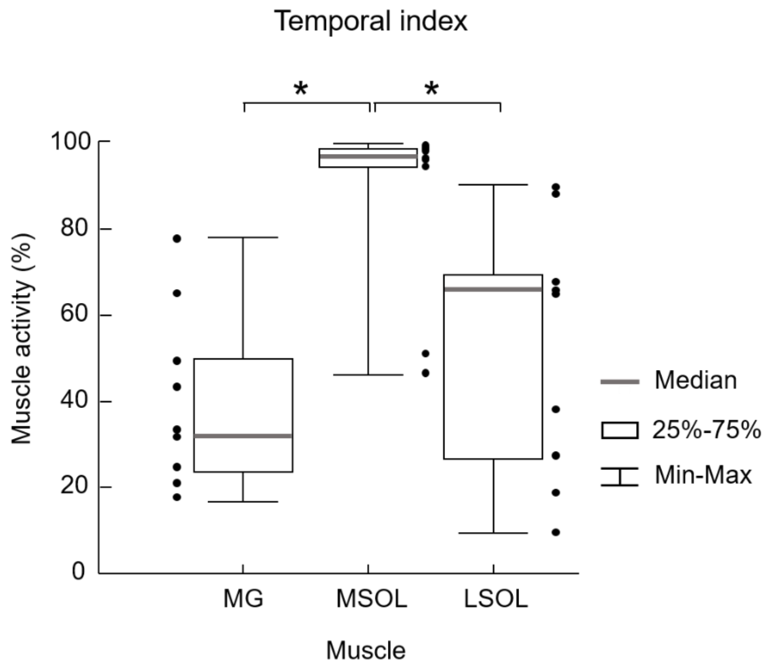

Figure 8: Box plot of temporal index of muscle activation. The temporal index was computed through the resultant ON-OFF signal of each muscle. Temporal index of MSOL was significantly higher than temporal index of MG and LSOL (* $p<0.05$).

Our results showed that, globally, calf muscles have a distinct temporal pattern of activation in standing balance. Specifically, SOL was activated almost continuously while MG was activated intermittently. Besides, different portions of SOL had a different duration of activation. These results support the notion of tonic innervation of SOL and phasic of the gastrocnemius for the control of postural sway in the standing posture [27,33]. Differences between the temporal activation of MG and SOL may depend on how far forward from the ankles an individual maintain his postural sway. On average, activity in the SOL muscle and very low modulation in the MG activity have been observed while subject sways closer to the ankles [33]. Once the subject sways forward, there is an increase of the number of motor units recruited and of the motor unit firing rates of MG and SOL [27,100]. Consequently, an agonist behavior in both muscles might occurs to correct forward sway events [33]. This implies that CNS seems to control both muscle independently, in which MG onset occurs to assist SOL muscle in the correction of forward sway. Therefore, CNS seems to adopt a control strategy based on intermittent synergistic effect of MG and SOL to maintain upright stance.

Furthermore, distinct proportions of muscle fiber type in the muscles of triceps surae could favor the continuous activation of SOL muscle while standing. It was observed that SOL contains a higher proportion of slow twitch fibers (70%) than MG (50%) [105] Thus, the recruitment of SOL MUs likely result in a muscular contraction characterized by low force generating capabilities and high fatigue resistance [106]. As low active ankle torques have been observed during upright stance [~5 Nm; [107]], SOL may be more suited to compensate the forward sway during standing balance. However, suppose that differences in the duration of activity between plantar flexors are accounted for differences in the fiber type composition between these muscles is questionable, since fiber type is highly adapted to physical demand [108] and, therefore, can be extremely variable among individuals [109].

Interestingly, distinct temporal patterns of activation were also observed between the medial and lateral portions of SOL muscle. Functional differences between the medial and lateral parts of SOL in the control of lateral sway could explain their distinct temporal patterns of activation. On one hand, the emergence of LSOL EMGs related to ankle eversion movement seems to indicate that LSOL also contribute in the compensation of lateral sway during standing balance [110]. On the other hand, the activation of medial muscles located at calf might contribute to ankle inversion movement, since MG activation together with other ankle inversors contribute to ankle lateral stabilization [111]. However, future studies are needed to confirm functional differences between MSOL and LSOL during standing balance.

These results suggest that CNS may control independently the compensatory activity of plantar flexors in standing balance. This selective activation of triceps surae may reflect an efficient control of standing balance, since CNS prioritizes an intermittent activation of MG and within SOL muscle instead of a continuous activation.

3.6 Conclusion

The results of this study indicate that global temporal activations of the muscles located at calf differ during standing balance. MSOL is continuously activated while LSOL and MG are intermittently activated during upright stance.

Chapter 4

The spatial distribution of ankle muscles activity discriminates aged from young subjects during standing [2]

4.1 Abstract

During standing, age-related differences in the activation of ankle muscles have been reported from surface electromyograms (EMGs) sampled locally. Given though activity seems to distribute unevenly within ankle muscles, the local sampling of surface EMGs may provide a biased view on how often and how much elderly and young individuals activate these muscles during standing. This study aimed therefore at sampling EMGs from multiple regions of individual ankle muscles to evaluate whether the distribution of muscle activity differs between aged and young subjects during standing. Thirteen young and eleven aged, healthy subjects were tested. Surface EMGs were sampled at multiple skin locations from tibialis anterior, soleus and medial and lateral gastrocnemius muscles while subjects stood at ease. The root mean square amplitude of EMGs was considered to estimate

[2] Scientific paper published in Frontiers in Human Neuroscience, 2017; Authors: Dos Anjos F, Pinto TP, Gazzoni M, Vieira TM.

the duration, the degree of activity and the size of the region where muscle activity was detected. Our main findings revealed the medial gastrocnemius was active for longer periods in aged (interquartile interval; 74.1–98.2%) than young (44.9–81.9%) individuals (P = 0.02). Similarly, while tibialis anterior was rarely active in young (0.7– 4.4%), in elderly subjects (2.6–82.5%) it was often recruited (P = 0.01). Moreover, EMGs with relatively higher amplitude were detected over a significantly wider proximo-distal region of medial gastrocnemius in aged (29.4–45.6%) than young (20.1–31.3%) subjects (P = 0.04). These results indicate the duration and the size of active muscle volume, as quantified from the spatial distribution of surface EMGs, may discriminate aged from young individuals during standing; elderlies seem to rely more heavily on the active loading of ankle muscles to control their standing posture than young individuals. Most importantly, current results suggest different conclusions on the active control of standing posture may be drawn depending on the skin location from where EMGs are collected, in particular for the medial gastrocnemius.

4.2 Introduction

Different mechanisms have been suggested to account for the control of human, standing posture. Of recent interest is the notion that bodily sways are arrested by timely, triggered bursts of calf muscles' activity [26–28]. The physiological mechanism underpinning timed activation in standing is controversial; some suggest there is an internal clock triggering calf muscles' activation [112] while others believe activation is event-triggered [29]. In spite of the potential sources accounting for such intermittent, postural activation, alternating periods of muscle activation with silencing seems advantageous. Periodically silencing postural muscles during standing may: (i) allow the nervous system to sense joint angles without interference from muscle contraction [36]; (ii) compensate for intrinsic, feedback delays [31]; (iii) reduce metabolic costs [29]; (iv) not increase postural instability [31]. In this view, the assessment of muscle activation may therefore reveal pivotal features of the control of standing posture, especially e.g., in persons with balance impairments.

Surface EMGs revealed, indeed, key differences in the activation of postural muscles with aging. From bipolar surface EMGs collected from the plantar flexor muscles, for example, previous studies reported that elderlies tend to stand with a more continuous and higher degree of activation than young individuals [12–14]. Moreover, while tibialis anterior is typically silent in young subjects during standing [33,34], in aged individuals it is often recruited [14]. From the point of view of muscle activation, it seems therefore standing becomes progressively more demanding with aging. A crucial question arising from previous studies is whether the greater, prolonged activation observed locally (i.e., with bipolar EMGs) in plantar and dorsal flexors of aged subjects reflects the activation of the muscle as a whole.

The spatial distribution of surface EMGs over the whole muscle rather than the local sampling of surface EMGs with bipolar electrodes seems to more likely provide a genuine indication on the duration and degree of muscle activity. EMGs with different amplitudes have indeed been observed when sampled from different regions of a single muscle [113,114], suggesting activity does not distribute uniformly within the muscle volume. This uneven distribution of activity has been often observed for the calf muscles and for a number of circumstances, including standing [20,23,115]. Methodologically, these results suggest the local sampling of surface EMGs may not provide a representative view of the degree and duration of calf muscles' activation. Physiologically, the differential distribution of activity within the calf muscles may indicate a key mechanism contributing to the control of muscle force and thus of the standing posture. It is therefore possible that differences in the activation of postural muscles with aging are more expressive than previously appreciated.

This study questions, for the first time, whether the distribution of muscle activity differs between aged and young subjects during standing. Differently from previous studies, here we use arrays of surface electrodes to sample EMGs from different regions of individual, ankle muscles. More specifically, from surface EMGs collected serially from ankle plantar and dorsal flexors we ask: do young and aged subjects activate an equal proportion of their muscles for a similar duration during standing? If aging is associated with greater muscular effort for standing control [12,13], we therefore expect to observe EMGs with greater amplitude for a longer duration and in a larger muscle region in aged than young individuals.

4.3 Material and Methods

4.3.1 Participants

Thirteen young male volunteers provided written informed consent before participating in the study (mean ± SD; age: 26 ± 3 years; body mass: 72.4 ± 10.1kg; height: 1.75 ± 0.06m) and 11 aged (70 ± 6years; 72.9 ± 12.5kg; 1.72 ± 0.08m). All participants were classified as minimally active according to the international physical activity questionnaire (IPAQ); short self-administered version [116]. All community-dwelling older adults lived independently. We decided to include participants without a sedentary lifestyle because physical inactivity may further broaden the between subjects variability often reported for stabilometric descriptors [117]. The experimental procedures considered in this study conformed to the Declaration of Helsinki and were approved by the Regional Ethics Committee (Commissione di Vigilanza, Servizio Sanitario Nazionale—Regione Piemonte—ASL 1—Torino, Italy). Volunteers did not report any balance impairments, neurological disorders, muscular injuries, or the intake of medications that could affect their standing balance at the occasion of experiments.

4.3.2 Experimental protocol

Participants were instructed to stand upright barefoot on a force-plate, with eyes open and arms alongside the body. They positioned their feet at a comfortable orientation and distance, while keeping heels and toes at the same position along the force plate anterior-posterior axis (Figure 9A). Prior to starting experiments, the contour of both feet was marked on the force plate to ensure participants would keep the same feet position throughout the standing tests.

Two standing tasks were applied. In the first task, participants were provided with visual feedback of their CoP position in the anterior-posterior axis and were instructed to keep it at 65% of the longitudinal size of their support base for 60s (Figure 9B). The size of the support base in the anterior-posterior axis was defined as the distance between the tip of the third metatarsal head and the tip of the calcaneus bone projected in the anterior-posterior direction (Figure 9A). The 65% figure corresponds roughly to 80% of the distance between the heels and the anterior limit of stability of healthy, young subjects [118]; this figure was selected to ensure a somewhat high degree of calf muscles active loading while not threatening stability, in particular for the aged individuals. All elderly subjects tested could stand with their CoP at the target value without losing balance. This postural task was considered for the normalization of EMGs, as described below. In the second task, volunteers were asked to stand at ease for 60s. Subjects were engaged in active conversation to ensure they would take their mind off the test and thus avoid any voluntary change in muscle activity [102]. The trial started over in the case gross body movements were noticed by the experimenter. The standing at ease task was applied three times, in accordance with previous evidence on excellent reliability of stabilometric descriptors [119]. Similarly, good-to excellent reliability has been recently reported for EMGs detected from the calf muscles in aged individuals during standing [120]. Five minute intervals were applied between trials and their order was randomized.

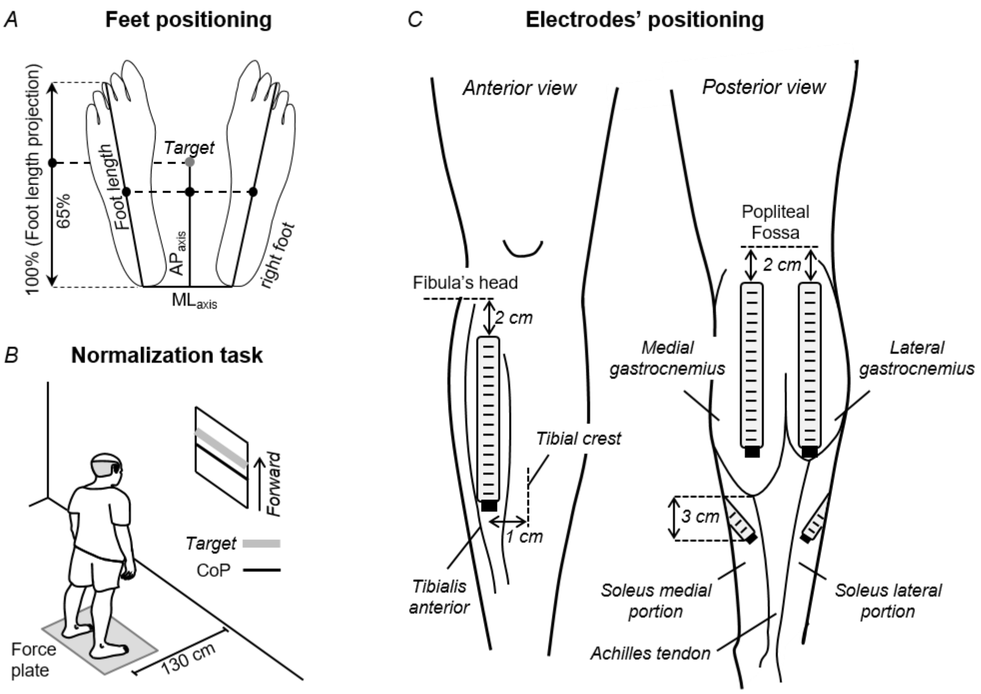

Figure 9: Standing protocol and feet and electrode positioning. (*A*) Schematic illustration showing the procedure considered to measure feet length. The midpoint between the tip of left and right calcaneus bones was considered as the origin of the reference system for center of pressure (CoP) and foot length measurements. The target value, corresponding to 65% of the foot length projected on the anterior-posterior axis (AP axis), was calculated and then considered for the normalization task shown in (*B*). In such task, participants were asked to keep their CoP position in the AP axis (black line) within 10% (±5%) of the target value (thick, gray line). (*C*) The positioning of electrode arrays is shown for the tibialis anterior muscle (left), for the medial and lateral gastrocnemius heads and for the soleus medial and lateral portions (right).

4.3.3 Electrodes' positioning

Linear arrays of surface electrodes were used to sample the distribution of ankle muscles' activity. Arrays were positioned over the plantar and dorsal flexors of both legs. Two arrays of 16 electrodes (10mm inter-electrode distance) were used to detect surface EMGs from the medial and lateral gastrocnemius muscles. The most proximal electrode was located 2cm distally to the popliteal fossa and arrays were aligned parallel to the longitudinal axis of each gastrocnemius' head (Figure 9C). Such positioning maximizes the representation of action potentials from muscle units residing in different, proximo-distal gastrocnemius' regions [17]. EMGs were sampled with one array of 16 electrodes (10mm inter-electrode distance) from the tibialis anterior muscle, aligned 1cm laterally and parallel to the tibial crest and with the most proximal electrode located 2cm distally to the fibula's head (Figure 9C). Given the in-depth pennate architecture of tibialis anterior, as for gastrocnemius, such positioning is expected to provide EMGs representative of different, proximo-distal fibers. Two arrays with four electrodes each (10mm inter-electrode distance) were used to sample EMGs from the soleus medial and lateral portions. For each soleus portion, arrays were aligned ∼45 degree outward to the line connecting the junction between gastrocnemius' heads and the calcaneus tip. The lower border of both the medial and lateral arrays was positioned 3cm distally to the medial gastrocnemius myotendinous junction (Figure 9C; [23]). Gastrocnemii junction and their myotendinous junction were identified with ultrasound imaging (cf. Supplementary Material in [101]). Arrays were positioned after cleaning the skin with abrasive paste.

4.3.4 Electromyographic and stabilometric recording

EMGs were recorded in single-differential derivation. All 51 single-differential EMGs were amplified by a between-individuals variable factor—5,000 or 10,000—to ensure the highest signal-to-noise ratio without saturation (10–750 Hz bandwidth amplifier; EMG-USB, OTBioelettronica and LISiN, Politecnico di Torino, Turin, Italy). CoP coordinates in the sagittal and frontal planes were computed from the ground reaction forces supplied by a piezoelectric force-plate (9286AA Kistler, Zurich, Switzerland). Reaction forces and surface EMGs were sampled synchronously at 2,048Hz using a 12-bit A/D converter (±2.5 V input dynamic range).

4.3.5 Quantifying muscle activity during standing

Raw surface EMGs were visually inspected. Whenever any channel in the array presented contact problems, likely due to high skin-electrode impedance, or massive power line interference, the corresponding channel was disregarded. Occurrences of low quality EMGs were infrequent (17 out of 1,224 EMGs detected in total) and were observed mainly in one channel per array. After controlling for signal quality, EMGs from tibialis anterior and gastrocnemius muscles were inspected for the identification of propagating potentials. Propagating potentials may be observed in the distal muscle region, where surface electrodes and muscle fibers may run in parallel direction. In such case, different electrodes sample from the same group of fibers rather than from different, proximo-distal fibers (cf. Figure 1 in [20]). Channels providing propagating potentials were therefore excluded from analysis; from 0 to 8 channels per array were excluded for the 24 participants tested.

After visual inspection, EMGs were band-pass filtered with a fourth order Butterworth filter (15–350Hz cutoff; zero lag, bidirectional filter). Then, the Root Mean Square (RMS) was computed over 40ms epochs [14], providing a total of 1,500 RMS values per channel. From these RMS values, three indices were considered to assess for how long, how diffusely and how much elderly and young activated their ankle muscles during standing.

Instants of activation were estimated by comparing the RMS values obtained during standing with the background activity. The background activity was defined from the RMS amplitude of EMGs collected with the ankle muscles at rest [14]. More specifically, for each channel in each array of electrodes we: (i) computed the RMS values over 40ms epochs for EMGs detected during 3s while participants were in supine position, providing a total of 75 RMS values; (ii) calculated the mean and the standard deviation for these RMS values; (iii) set the threshold defining the background noise level as the mean value plus three standard deviations; (iv) assigned Active or Inactive state to RMS samples respectively exceeding or not exceeding the background threshold. This procedure provided a series of Active-Inactive state values per channel. Given the EMGs detected by consecutive electrodes in the array sample from different fibers along the muscle proximo-distal axis, concurrent Active-Inactive events were often not observed between channels (Figure 10A). For this reason, to provide a global indication on the duration of muscle activity during standing, the muscle was deemed active whenever an Active state was observed across channels (cf. shaded areas in Figures 10A, B). The global series of Active-Inactive states indicates how long the ankle muscles were active throughout standing, regardless of where activity was observed in the muscle.

Based on the instants of activation, the spatial distribution and the intensity of muscle activity were computed. First, the number of channels detecting surface EMGs with RMS amplitude greater than 70% of the highest RMS amplitude in the array was identified. The 70% amplitude threshold was selected because it has been shown to provide a robust identification of channels located over active fibers within muscles pennate in the depth direction [121]. The number of segmented channels multiplied by the inter-electrode distance was then normalized with respect to the muscle length (see below). Second, the degree of muscle activity was estimated by averaging the RMS values across all segmented channels for each muscle tested. This average RMS amplitude was then normalized with respect to the RMS amplitude averaged across channels identified during the normalization, standing task (Figure 9B), to compensate for the effect of anatomical differences between participants on the surface EMGs [122]. It should be noted here the

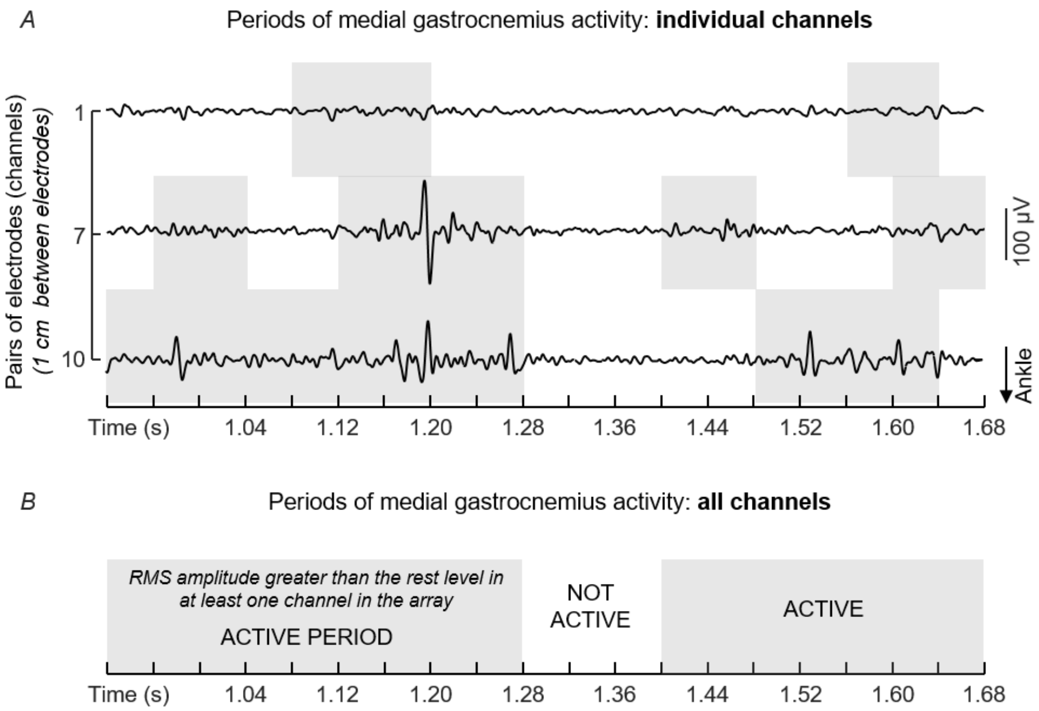

Figure 10: Computation of instants of muscle activation. (*A*) Shows examples of single-differential EMGs detected by channels 1, 7, and 10 from the medial gastrocnemius muscle of a single participant. Note action potentials do not appear with equally high amplitude in the three channels. The muscle was considered active whenever for any given instants the RMS amplitude exceeded the background activity in at least one channel. Light gray rectangles in the grid denote periods within which the RMS amplitude of EMGs in each channel exceeded the background activity computed with the muscle at rest (see text). The resulting series of instants of muscle activity for this representative example is illustrated with a succession of white (Inactive state; muscle at rest) and gray (Active state; muscle active) rectangles (*B*).

normalization of EMGs collected in a given condition with respect to that collected in a reference condition compensates for the effect of inter-individual differences on their amplitude though not on the spatial distribution of their amplitude. Finally, the size of the active region in the proximo-distal direction and the degree of activity were considered for analysis whenever any given muscle was active for at least 10% of the total, standing duration. Following previous evidence, occurrences of such sporadic activity was regarded as of marginal relevance for the control of standing posture [26]. Given we observed medial gastrocnemius was active for different durations (~20%) between legs in both aged and young subjects, the electromyographic indices and architectural muscle parameters (see below) computed from the subject's leg active for longer durations (left side for 10 young and 8 elderly subjects) were used for comparisons between groups.

4.3.6 Measurements of ankle muscles' length and subcutaneous thickness

Parasagittal images from tibialis anterior and gastrocnemii were taken with a linear, ultrasound probe (10MHz, 4cm length; Echo Blaster 128, Telemed Ltd., Vilnius, Lithuania), with participants lying comfortably on a padded bed in prone position. The participants' feet were positioned out of the padded bad. Initially, the myotendinous junction was identified and marked on the skin. The length of tibialis anterior was then quantified as the distance between its myotendinous junction and the head of the fibula, whereas the length of each gastrocnemius head was defined as the shortest distance between the myoetendinous junction and the popliteal fossa. Subcutaneous thickness and pennation angle were further quantified to assist in the interpretation of potential proximo-distal differences in the distribution of activity with aging. Subcutaneous thickness was computed as the distance between the skin/fat layer over the muscle and its superficial aponeurosis from parasagittal images obtained with the probe centered halfway the muscle length. Thickness measurements were taken from the central region of the proximal, central and distal thirds of the ultrasound images and then averaged, providing a representative indication on the general subcutaneous thickness per subject [123]. Finally, the pennation angle was estimated as the angle between a clearly visible fascicle in the image and the muscle deep aponeurosis [124]. Thickness and pennation angle values were obtained with the muscle at rest.

4.3.7 Quantifying the CoP sway area

The CoP sway area was considered to assess how largely young and elderly individuals swayed during the whole standing tests. The overall size of postural sways was estimated from the elliptic area conveying almost 85% of the total CoP samples during standing [125]. The CoP elliptic area, as well as the EMG descriptors, were averaged across the three standing trials and considered for between-group comparisons. CoP data was 50Hz low-pass filtered with a second order Butterworth filter to remove high-frequency noise.

4.3.8 Statistical analysis

Non-parametric statistics were applied to compare the distribution of ankle muscles' activity between young and aged individuals, after ensuring the data distribution was not Gaussian (Shapiro-Wilk's W-test, $P < 0.03$ in all cases). The Mann-Whitney U-test was used to verify whether, during standing and for each muscle independently, the duration of the active period, the degree of activity, the relative size of the active region and the CoP elliptic area were different between groups. The same statistics was applied to assess regional differences in activity within soleus, by comparing the normalized RMS amplitude obtained for the muscle medial and lateral aspects. The level of statistical significance was set at 5% and data were reported using non-parametric, descriptive statistics.

4.4 Results

Representative examples of variations in ankle muscles' activity during standing

Different ankle muscles were activated differently when young and aged individuals stood at ease. Descriptive considerations on these differences are summarized in this section based on the data shown in Figure 11 for one young and one aged, representative participant. Lateral gastrocnemius was active for 6 and 17% of the whole standing trial for the young and aged subjects respectively. For the tibialis anterior muscle, the amplitude of EMGs remained below the background activity for the young subject during the whole standing duration while, for the elderly, bursts of activity were observed and provided an active duration of 36%. Conversely, EMGs with remarkably high amplitude in the soleus medial and lateral portions were observed consistently for both subjects (cf. shaded areas in Figure 11).

The medial gastrocnemius showed a somewhat different pattern of activity when compared with the other ankle muscles. Differently from lateral gastrocnemius, soleus and tibialis anterior, medial gastrocnemius was not completely silenced, was not activated almost continuously and did not show sporadic bursts of activity. For both participants whose data is shown in Figure 11, alternate periods of medial gastrocnemius activation and silencing were observed consistently during standing (Figure 12A). When considering the distribution of activity within medial gastrocnemius, EMGs with relatively high amplitude were detected by six channels in the young and by nine channels in the aged subject (cf. gray circles in Figure 12B).

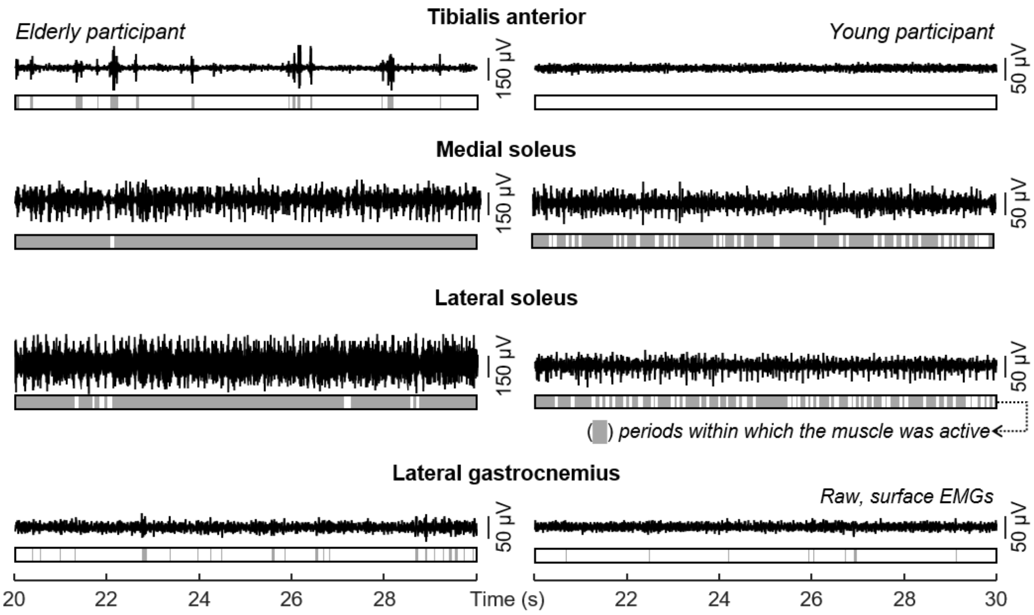

Figure 11: Modulations in ankle muscles' activity during standing. Single-differential EMGs collected for a representative channel from the tibialis anterior, the soleus medial and lateral portions and from the lateral gastrocnemius of a representative young and elderly participant are shown for 10s of standing at ease. Light gray areas in the bars indicate periods within which the RMS amplitude exceeded the background activity (i.e., Active state).

Ankle muscles' activation in elderly and young individuals during standing

When considering group data, differences in the duration of ankle muscles' activity were observed between elderly and young subjects. The Mann-Whitney U-test revealed the medial gastrocnemius was activated for longer periods in aged (median, interquartile interval; 81.2, 74.1–98.2%) than young (58.8, 44.9–81.9%) individuals (Figure 13A; P = 0.02; N = 24; 13 young × 11 aged subjects). Similarly, notwithstanding the marked variability in the duration of active periods for the elderly, aged participants activated their tibialis anterior muscle during standing for a significantly longer duration than young subjects (*P* = 0.01; N = 24); tibialis anterior was rarely active in young (1.3, 0.7–4.4%) though not in the elderly (29.1, 2.6–82.5%). No group differences were observed in the duration of lateral gastrocnemius (*P* = 0.07) and soleus medial (*P* = 0.90) and lateral (*P* = 0.26) portions.

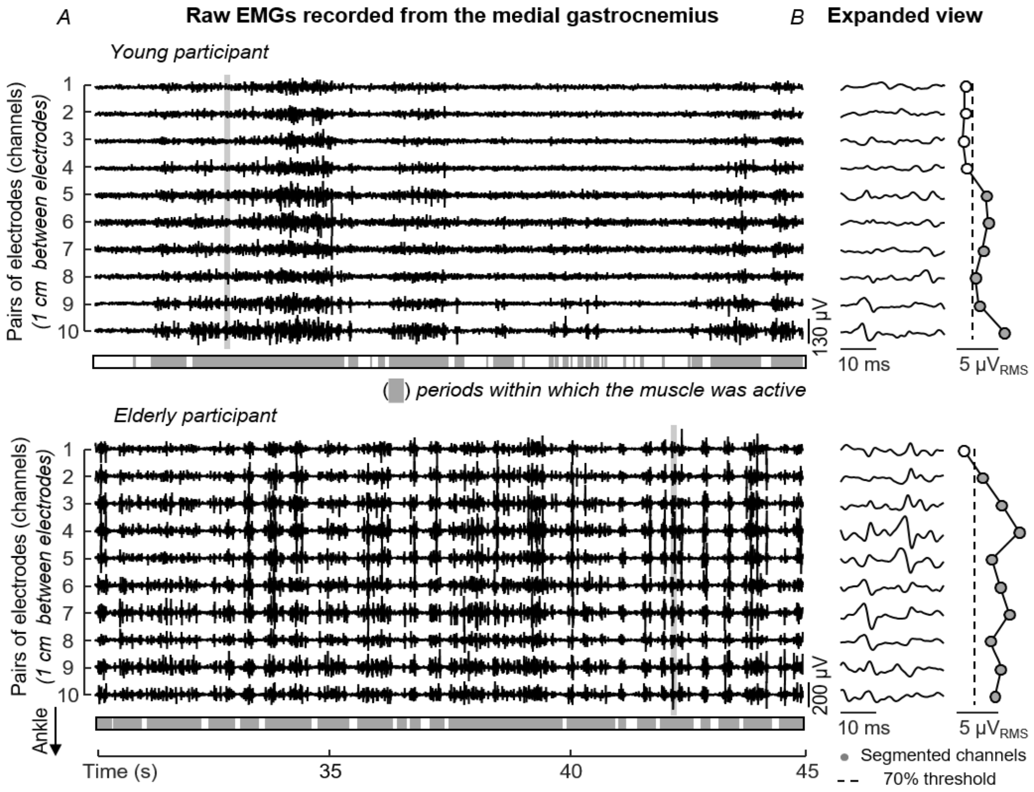

Figure 12: Duration and distribution of medial gastrocnemius activity during standing. Raw EMGs detected by channels located over the gastrocnemius superficial aponeurosis, from channel 1 to 10, are shown in (*A*) for a young and elderly representative participant. Light gray areas indicate periods within which the RMS amplitude exceeded the background activity. (*B*) Shows an expanded view of all EMGs. Note different action potentials appear in different channels for each subject. Note also there is no delay between potentials detected by consecutive channels (e.g., channels 1–6 for the aged participant). The distribution of RMS amplitude across channels is represented from circles, with gray circles indicating the segmented channels; that is, channels detecting largest EMGs in the array (see text).

The degree and the distribution of activity within ankle plantar flexors varied differently between groups. The RMS amplitude of EMGs collected from the medial gastrocnemius and from the medial and lateral soleus portions did not differ between elderly and young (Figure 13B; $P > 0.40$ in all cases; N = 24). No differences were observed between the RMS amplitude for EMGs collected medially and laterally from soleus muscle, both for young and elderly ($P > 0.70$ in both cases). In contrast, for medial gastrocnemius, EMGs with relatively higher amplitude were detected over a significantly wider proximo-distal region in aged (32.5, 29.4–45.6%) than young (29.8, 20.1–31.3%) subjects (Figure 13C; $P = 0.04$; N = 24). Given the lateral gastrocnemius and tibialis anterior of young individuals were active for a somewhat short period during the whole standing tests (less than 10% on average; Figure 13A), these muscles were disregarded from further consideration.

Medial gastrocnemius' subcutaneous thickness and pennation angle

No differences in subcutaneous thickness were observed between groups ($P = 0.58$). Thickness values ranged from 1.6 to 2.7mm for young and from 1.6 to 3.5 for aged participants. Similarly, the pennation angle did not differ significantly ($P = 0.37$) between young (range: 19.6–24.0 degrees) and aged subjects (17.2–25.0 degrees).

Differences in CoP sway area with age

While aged participants stood at ease, their CoP occupied an area roughly twice (median, interquartile interval; 4.4, 3.8–9.6cm2) as large as that (2.3, 1.6–4.5 cm2) confining the CoP of young participants ($P = 0.052$).

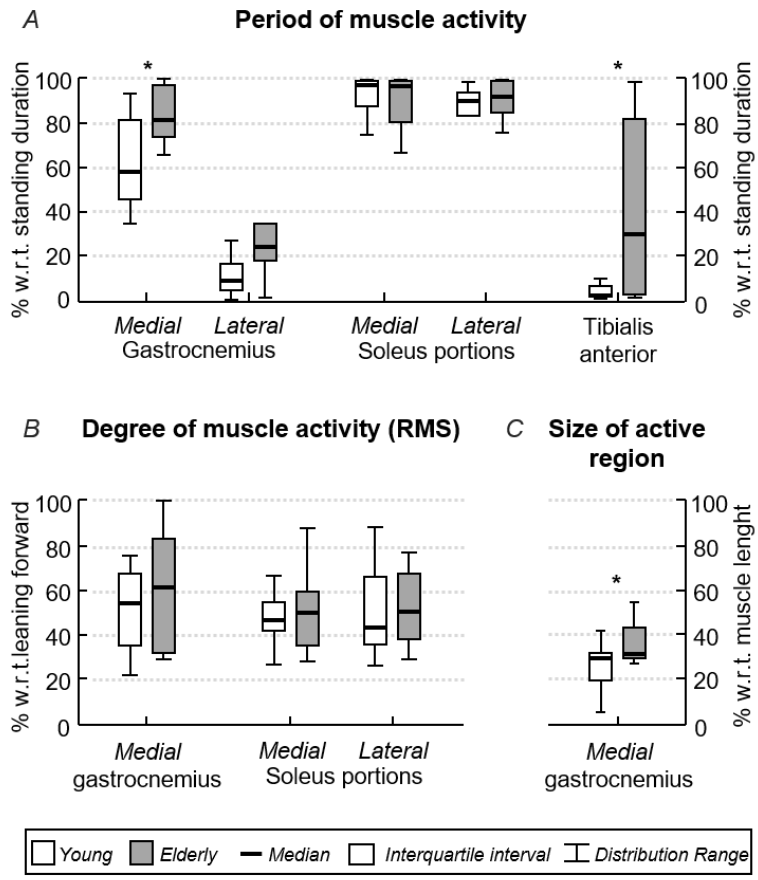

Figure 13: Age-related differences in the ankle muscles' activity during standing. Boxplots in (*A*) show the relative period within which, during standing, young (white boxes) and elderly (gray boxes) individuals activated their ankle muscles. Group data in (*B*) corresponds to the normalized, RMS amplitude obtained for both groups from the medial gastrocnemius and from the medial and lateral soleus portions. The proximo-distal size of medial gastrocnemius region over which largest EMGs were detected in both groups is shown in (*C*). Asterisks indicate significant differences between groups ($P < 0.05$).

4.5 Discussion

In this study we used arrays of electrodes to investigate whether the temporal and spatial distributions of ankle muscles' activity differ between elderly and young individuals during standing. We hypothesized that elderlies would present greater EMGs, distributed over a larger muscle region and for a longer duration than young individuals. Surface EMGs from different regions of the ankle muscles were collected while subjects stood at ease to test this hypothesis. Our key results indicate that during standing: (i) tibialis anterior and medial gastrocnemius muscles were active for a longer duration in aged than young subjects; (ii) a greater proportion of medial gastrocnemius volume was active in elderlies. Collectively, these results indicate elderlies rely more heavily on the active loading of ankle muscles to control their standing posture than young individuals.

Preliminary, methodological considerations on surface EMG detection

Differently from previous studies, here we sampled surface EMGs with arrays of electrodes (Figure 9C). Our decision to sample activity from multiple locations of individual ankle muscles was motivated by recent evidence suggesting different muscle regions may be activated differently [113,114], in particular for pennate muscles [17,20,22,115]; EMGs sampled locally (i.e., with a single pair of shortly spaced electrodes) may not provide a representative view of muscle activation. Indeed, as shown in Figure 10, EMGs in different locations may provide different estimations for the duration of gastrocnemius activity. Considering EMGs were represented with different amplitude in different muscle regions (Figures 10, 12), the duration of muscle activity would have been likely underestimated if we had not considered EMGs detected at different skin regions (cf. shaded areas for individual and all channels in Figure 10). Similarly, biased estimations on the degree of each calf muscles' activity would have been obtained if we had considered the RMS amplitude of EMGs detected locally. Depending on where EMGs were detected from, in particular for young individuals (Figure 12), their corresponding RMS values would provide either under- (white circles in Figure 12B) or over (gray circles in Figure 12B) estimates of the degree of activity in the whole muscle. Through arrays of surface electrodes, we were able to obtain estimates of the duration and degree of activity presumably more representative of individual ankle muscles' volumes than previously appreciated. As discussed below, new insights have been gained into age-related differences in postural activation from such a high-density, surface EMG approach.

Are the ankle muscles activated for a similar duration in aged and young subjects?

The differences between aged and young individuals reported here were muscle dependent. When considering the duration of activity for lateral gastrocnemius and soleus muscles, statistic differences were not observed between groups. Both groups recruited lateral gastrocnemius for less than ∼20% of the total, standing duration (Figures 11, 13A). The soleus muscle, on the other hand, was activated for periods roughly longer than 80% of standing. These results corroborate previous findings on the duration of calf muscles' activity, extensively reported for young subjects. Indeed, while the absence of activity in lateral gastrocnemius has been observed in some subjects [26,34], it seems well-documented that soleus is recruited at almost all time during standing [14,26]. Even though the temporal activation of postural muscles is not as commonly assessed for elderlies as it is for young individuals, Laughton et al. [14] observed age similarities for the duration of soleus activity during standing. Extending the observation of Laughton et al. [14], our current results suggest the duration of activity of the lateral gastrocnemius and soleus regions accessed by our surface electrodes is unlikely sensitive to aging.

Age differences emerged however in the duration of medial gastrocnemius and tibialis anterior activity during standing. Aged individuals activated their medial gastrocnemius muscle for a period ∼20% longer, on average, than their young counterparts (Figure 13A). Similarly, tibialis anterior was active for a longer (∼30%) period in aged than in young subjects (Figure 13A). The median duration of tibialis anterior activity observed here, both for young (∼2%) and elderlies (∼30%), is well in agreement with the duration of activity periods reported by Laughton et al. [14] for this muscle. From a first inspection of results presented in Figure 13A, we feel inclined to consider the co-activation mechanism as responsible for the age differences in the timing of plantar and dorsal flexors' activation during standing. More specifically, while young subjects predominantly activated their plantar flexors, elderlies activated both plantar and dorsal flexor muscles during standing (Figures 11, 12A, 13A). In spite of controversies on whether co-activation may be detrimental or may compensate for poor control of posture, co-activation is often associated with increased postural sways [12,13,126]. Corroborating this common view, in the elderlies, we observed the CoP was confined within an elliptic area almost twice as large as that confining the CoP of young subjects. Regardless of the mechanisms underpinning age-differences in posture control, current results here suggest elderlies activate their ankle muscles for longer durations during standing than young subjects.

Do elderly and young recruit their calf muscles to a similar extent during standing?

Two notes on our estimates of the degree of activity are necessary before interpreting results. Young subjects activated tibialis anterior and lateral gastrocnemius somewhat rarely (Figures 11, 13A). We therefore disregarded both muscles from age comparisons. A second observation concerns how we conceived muscle activity. While the amplitude of EMGs is traditionally considered to assess the degree of muscle activity, here we considered both the amplitude and the size of skin region where EMGs with relatively high amplitude were sampled (Figures 10, 12). As argued below, the extent to which elderly and young activate their calf muscles depends both on the amplitude and on the amplitude distribution of surface EMGs.

The proximo-distal distribution of the amplitude of medial gastrocnemius EMGs distinguished aged from young individuals during standing. While young subjects presented EMGs with relatively high amplitude in the gastrocnemius distal region, corroborating previous findings [20], EMGs with similarly high amplitude were sampled from a larger, proximo-distal muscle region in aged individuals (c.f. gray circles in Figures 12B, 13C). Different sources could have accounted for a somewhat extensive distribution of EMG amplitude in the elderlies. A first issue to consider is the potential difference in muscle architecture between groups. Both subcutaneous thickness and pennation angle have shown to affect dramatically the amplitude distribution of surface EMGs; thicker subcutaneous tissue and smaller pennation angle result both in more spatially diffused surface EMGs [122,127]. Current results however did not indicate significant, age differences in gastrocnemius architecture. The lack of anatomical differences between groups leads us to consider the possibility that elderlies activated a larger gastrocnemius volume than young subjects during standing. In spite of recent controversies [128,129], it seems well-accepted that, for the gastrocnemius muscle, EMGs sampled in different proximo-distal regions reflect the activity of different muscle fibers [17]. The amplitude of EMGs detected distally and proximally, for example, is associated with the number of active fibers in the muscle distal and proximal regions respectively [127]. Given the mean amplitude of EMGs sampled from medial gastrocnemius was similar between groups (Figure 13B), the wider skin region from where EMGs with high amplitude were detected (Figure 13C) suggests a relatively greater proportion of muscle fibers may have been recruited in the elderlies.

Age differences in the distribution of activity for soleus were not as clear as for gastrocnemius. Since soleus is largely covered by the gastrocnemius muscles, spatial differences in soleus activation were assessed by comparing the amplitude of EMGs detected laterally and medially (Figure 9C). Our decision to assess both regions was based on previous evidence showing medio-lateral differences in soleus activation [22,23]. Given we were here interested in assessing the degree of muscle activity during standing, sampling EMGs unilaterally could provide a biased view on the actual degree of soleus, postural activity. Indeed, even though results in Figure 13B indicate EMGs with equal amplitude were detected medio-laterally for both age groups, stating young and aged subjects activate a similar proportion of their soleus muscle in standing is potentially fairly speculative. Considerations on the relevance of spatial differences in soleus activation with age are therefore conditioned to the possibility of sampling EMGs from a greater soleus region (with e.g., intramuscular electrodes) in future investigations.

Future perspectives and limitations

Spatial differences in the amplitude of EMGs collected from elderlies and young during standing have methodological and physiological implications. A first crucial point to consider is the localized representation of surface EMG during standing. Here we show that, depending on where a single bipolar EMG is collected from the gastrocnemius muscle, different conclusions maybe drawn on age differences during standing. The local sampling of muscle activity provided by bipolar surface EMGs may indeed contribute to explaining current disparities observed in the literature. While some studies did not report age differences in the degree of plantar flexors' activity during standing at ease [14,130], others have however documented a significantly higher degree of plantar flexor activation with aging [12,13]. Anticipating the differences between studies were due to inappropriate EMG sampling is a statement we strongly discourage. On the other hand, though, our current results suggest that different interpretations may emerge from EMGs detected locally during standing. Of more physiological, applied interest, is the suggestion that elderlies tend to stand with a greater degree of muscle effort than young subjects (Figures 13A, C). According to recent evidence, standing with minimal muscular effort while affording some degree of bodily sways may be advantageous; it may reduce the metabolic cost of standing [29], eliminate motor noise [36] and compensate for delays in the feedback loop [31]. While acknowledging the value of different protocols so far devised for the balance training [60,131], here we suggest that learning to efficiently activate dorsal and plantar flexors during standing could be beneficial for the control of standing posture in aged individuals.

Some additional considerations on the results presented here are necessary. Both young and aged subjects showed a somewhat large variability in the duration of ankle muscles activity (Figure 13A). A first possible explanation to such inter-individual variability is the difference in ankle stiffness between subjects [6]. Additionally, the inter-individual variability in muscle activation during standing, in particular the duration of tibialis anterior activation in the elderly, may be due to different neural sources. With aging, an assortment of impairments arising at the peripheral and central levels of the nervous system may develop, each impacting detrimentally and differently on the control of the standing posture [132]. By testing subjects without a sedentary lifestyle we expect to have limited the repertoire of posture-related impairments affecting present results. A final consideration regards whether crosstalk could have affected our results. We believe this possibility is unlikely. First because action potentials were represented locally in the surface EMGs. For example, the fact that action potentials detected from a given gastrocnemius region did not appear in neighbor channels (cf. Figure 10 and expanded view of EMGs in Figure 12) suggest any crosstalk from muscles other than gastrocnemius was relatively marginal. Moreover, the duration of activity of both muscles would be the same if crosstalk from soleus had contributed substantially to EMGs collected from gastrocnemius. In spite of these considerations, our results show the duration and the region of muscle activity, as quantified from surface EMGs, discriminate well-aged from young individuals during standing.

Chapter 5

Does the plantar flexors activity differ between lower limbs during standing?

5.1 Abstract

Inferences on the active contribution of plantar flexors to the stabilisation of human standing posture have been drawn from surface electromyograms (EMGs). Surface EMGs were however often detected unilaterally, presuming the myoelectric activity from muscles in a single leg reflects the pattern of muscle activation in both legs. In this study we question whether surface EMGs detected from plantar flexor muscles in both legs provide equal estimates of the duration of activity. Arrays of surface electrodes were used to collect EMGs from gastrocnemius and soleus muscles while twelve, young male participants stood at ease for 60 s. Muscles in each leg were deemed active whenever the Root Mean Square amplitude of EMGs (40ms epochs) detected by any channel in the arrays exceeded the noise level, defined from EMGs detected during rest. The Chi-Square statistics revealed significant differences in the relative number of active periods for both muscles in the majority of participants tested, ranging from 2% to 65% (χ^2>17.90; P<0.01). Pearson correlation analysis indicated side differences in the duration of gastrocnemius though not soleus activity were associated with the centre of pressure

mean, lateral position (R=0.60; P=0.035). These results suggest therefore that surface EMGs may provide different estimates of the timing of plantar flexors' activity if collected unilaterally during standing and that asymmetric activation may be not necessarily associated with weight distribution between limbs. Depending on the body side from which EMGs are collected, the active contribution of plantar flexors to standing stabilization may be either under- or over-valued.

5.2 Introduction

Insights into the neuromuscular mechanisms underpinning the control of human standing posture have been gained from surface EMGs. It seems well established, for example, that medial gastrocnemius is activated intermittently and soleus is activated continuously during standing [26,33] . An apparently, equally well accepted notion in the literature is the neural stiffening of the ankle joint in unstable circumstances, as suggested by the increased degree of co-activation of plantar and dorsal flexors [133,134]. Moreover, different mechanisms for standing control have been implicated from calf muscles' surface EMGs [2,28,135,136]. While all these pieces of evidence substantiate the potential relevance of surface electromyography, inferences on the neuromuscular determinants of posture control have been often drawn from EMGs collected from calf muscles in either left or right leg.

Side differences in calf muscles' activation have been however observed from surface EMGs detected in different standing conditions. Uneven distribution of body weight, muscle fatigue, disturbances in proprioceptive or vestibular inputs and proximity of the center of gravity vertical projection to the ankle joint are some factors possibly associated with asymmetric activation of ankle plantar flexors [137–140]. While asymmetric activation of plantar flexors may be well expected during prolonged standing, when e.g. occurrences of shifts in body weight between limbs are frequent [141], the differential activation of plantar flexors between legs seems controversial during quiet standing. For example, although Masani et al. [34] reported the left and right plantar flexors are activated equally during standing others detected plantar flexors' EMGs with different amplitudes between legs [139,142]. On top of these disparities, recent evidence has shown surface EMGs sampled locally from gastrocnemius unlikely represent the whole muscle volume [17,20]. It seems therefore relevant to ask whether inferences on the activation of plantar flexors may be drawn from EMGs collected unilaterally while subjects stand at ease.

In this study we investigate whether left and right plantar flexors are elicited for equal durations during standing. Instances of muscle activation are estimated from surface EMGs sampled with arrays of electrodes from different regions of the gastrocnemius and soleus muscles, ensuring action potentials represented in different muscle regions would contribute to estimating periods of muscle activity [143]. If plantar flexors of both legs are activated equally during standing, we expect to observe EMGs with relatively high amplitude for similar durations between limbs. To our knowledge this is the first study to systematically evaluate the bilateral representation of plantar flexors' myoelectric activity during standing.

5.3 Methods

5.3.1 Participants

Twelve young male volunteers (range: 24-34 years; 60-90 kg; 1.70-1.87 m) participated in this study after providing written, informed consent. Experimental procedures conformed to the *Declaration of Helsinki* and were approved by the Local Ethics Committee (Commissione di Vigilanza, Servizio Sanitario Nazionale—Regione Piemonte—ASL1—Torino, Italy).

5.3.2 Experimental protocol

Two different tasks were applied. In the first task, subjects were instructed to relax their muscles completely while lying supine on a padded bed. Surface EMGs collected at this condition were considered to set the background, noise level as indicated below. In the second task subjects stood on a force plate for 60 s, with eyes open, arms alongside the body and feet in a comfortable position. Their heels and toes were aligned parallel to the force plate lateral axis (Figure 14). The feet contour was marked on the force plate to ensure participants would keep the same stance throughout experiments. Subjects were engaged in active conversation to suppress any voluntary control of calf muscles' activity during standing [102]. The second task was applied three times, with 2 min intervals in-between.

5.3.3 Signal recordings

Single-differential EMGs were collected from the soleus and medial gastrocnemius muscles of both legs with linear arrays of silver-bar, surface electrodes (1x10mm; 10mm inter-electrode distance). Arrays of 16 electrodes were positioned over medial gastrocnemius muscles, aligned parallel to the muscle longitudinal axis and with the most proximal electrode located 2 cm distally to the popliteal fossa (Figure

14A). Two shorter arrays with four electrodes each were used to sample EMGs from soleus; arrays were oriented ~45° outward to the line connecting the junction between the gastrocnemii and the calcaneus and the distal electrode was located 3 cm distally to the medial gastrocnemius myotendinous junction (Figure 14A). Ultrasound imaging (linear probe; 10 MHz; Echoblaster 128, Telemed Ltd., Vilnius, Lithuania) was used to identify gastrocnemii and myotendinous junctions (cf. supplementary material in [101]). Electrodes were positioned after cleaning the skin with abrasive paste.

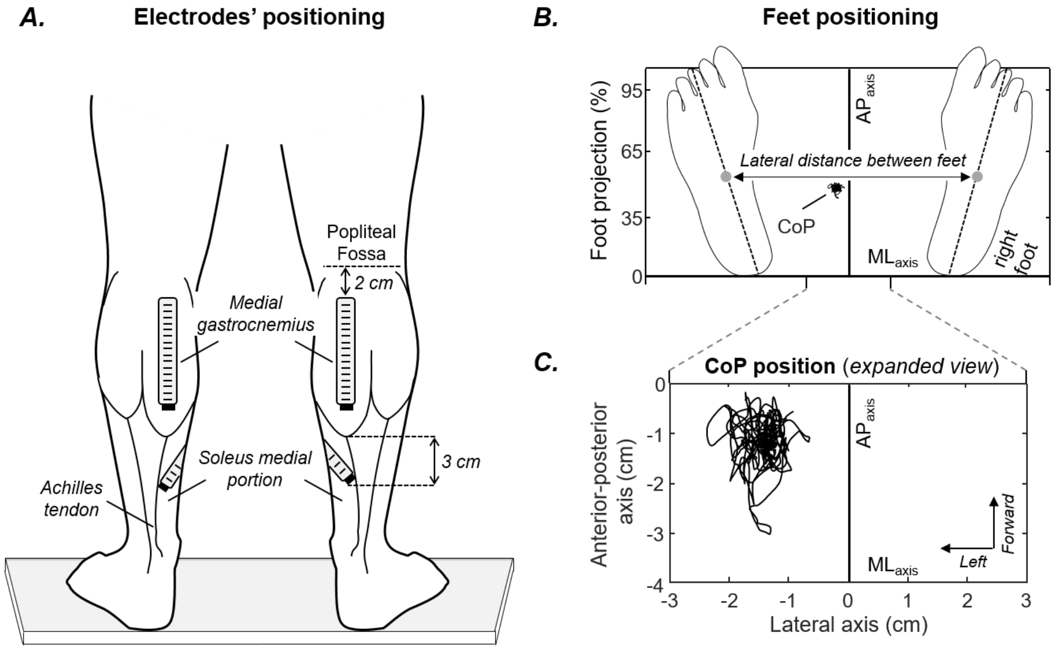

Figure 14: Electrodes and feet positioning. *A*, shows the positioning of electrode arrays for the medial gastrocnemius and soleus muscles. *B*, a schematic illustration of feet positioning is shown. Foot length was calculated as the distance between the tip of the third metatarsal head and the calcaneus bone. The distance between the centers of the length of each foot was considered to define the lateral distance between feet and thus the anterior-posterior axis. *C*, an expanded view of CoP shown in *B*.

EMGs were amplified by a variable factor across subjects (5,000-10,000) to maximize the signal-to-noise ratio (10-750Hz bandwidth amplifier; EMG-USB, OTBioelettronica and LISiN, Politecnico di Torino, Turin, Italy). Signals were digitized at 2048 Hz using a 12-bit A/D converter (±2.5V dynamic range). Ground reaction forces supplied by a piezoelectric force-plate (9286AA Kistler, Zurich, Switzerland) were sampled synchronously with EMGs.

5.3.4. Assessment of muscle activity

Initially, raw EMGs were visually inspected for the identification of channels with contact problems or massive power line interference; 13 out of 432 channels were discarded. Moreover, specifically for gastrocnemius, the most distal channels in the array detecting propagating potentials were excluded. The presence of propagating potentials indicates the distal channels provide redundant information on muscle activation, given they sample from the same rather than from different gastrocnemius fibres (cf. Figure 1 in [20]). After visual inspection, EMGs were band-pass filtered (15–350Hz cut-off; 4th order Butterworth bidirectional filter). Root Mean Square (RMS) amplitude was then computed over 40ms epochs [14], providing a total of 1,500 RMS values per channel.

The duration of muscle activity was estimated from RMS values. First, background activity was defined from RMS values calculated for EMGs detected with the plantar flexors at rest; background level was set as the mean plus three standard deviations calculated over 3s of rest (40ms epochs;[14]). Background level was defined separately for each channel in each array of electrodes. Whenever the RMS amplitude of any channel in a given array exceeded the background level, the corresponding muscle was deemed active during standing. This procedure provided a series of *active* instances, indicating the proportion of the standing test during which the activity of each muscle in each limb exceeded the background level (cf. Figure 2 in [143]). Finally, the median number of *active* periods across the three standing tests was considered for analysis.

5.3.5 Calculation of CoP lateral position

CoP position was computed from the vertical, ground reaction forces and then low-pass filtered (5 Hz cut-off; 2nd order Butterworth filter). CoP coordinates in the frontal plane were averaged across the whole standing duration and across the three standing tests. This mean CoP position was then: i) referred to the anterior-posterior axis passing midway through the lateral distance between feet (Figure 14B), with negative values indicating CoP was on average closer to the left foot; ii) normalized with respect to the lateral distance between feet, compensating for the different, comfortable stances adopted by participants.

5.3.6 Statistical Analysis

Contingence tables were created from the median number of active instances, separately for each muscle and subject. The Chi-square (χ^2) test was then applied to test for whether the proportion of *active* periods between limbs was similar during standing [144]. After ensuring the data distribution was Gaussian (Shapiro-Wilk's W test, P>0.23 in all cases), Pearson correlation was applied to verify whether side-differences in the duration of activity (i.e., right/left ratio of the number of active periods) were associated with the CoP mean position in the frontal plane.

5.4 Results

Side-differences in plantar flexors' activity

Activation periods obtained from surface EMGs detected from different muscle regions were not the same. As shown for a representative participant (5s epoch; Fig. 15A), EMGs with relatively high amplitude were predominantly detected from the distal region of the right gastrocnemius muscle. Close inspection of an expanded view of these raw EMGs further indicates that action potentials of different motor units were detected from different regions, resulting in the identification of different periods of activity across channels. Our procedure for estimating the duration of muscle activity was not sensitive however to regional differences in EMG amplitude; regardless of where action potentials were detected from the muscle they were considered to estimate periods of activity (cf. grey rectangles in Fig. 15B right panel).

Statistically significant side differences in the duration of activity were observed for at least 10 out of 12 participants tested during standing (Fig. 16). The absolute right-left difference in the duration of activity ranged from 3.7% to 65.3% for the gastrocnemius muscle (Fig. 16A; χ^2>33.35; P<0.01; N=10 subjects). For the soleus muscle, this difference ranged from 2.0% to 37.2% (Fig. 3B; χ^2>17.90; P<0.01; N=11 subjects). Differences in the duration of activity were not observed consistently for the same side and muscle; two and four participants activated respectively more frequently the gastrocnemius and soleus muscles of the right leg (cf. circles and squares in Fig. 16). Although participants five and six activated the left and right gastrocnemius for a similar, relative duration (~50%; Fig. 16A), these muscles were concurrently active during less than 30% of the standing test (cf. grey rectangles in Fig. 16). Finally, for all participants, regardless of the leg considered, soleus was active for either a longer (χ^2>4.19 and *P*<0.04 for 23 legs) or similar (χ^2=1.15 and *P*=0.28 for the right leg of the subject five) duration when compared to gastrocnemius.

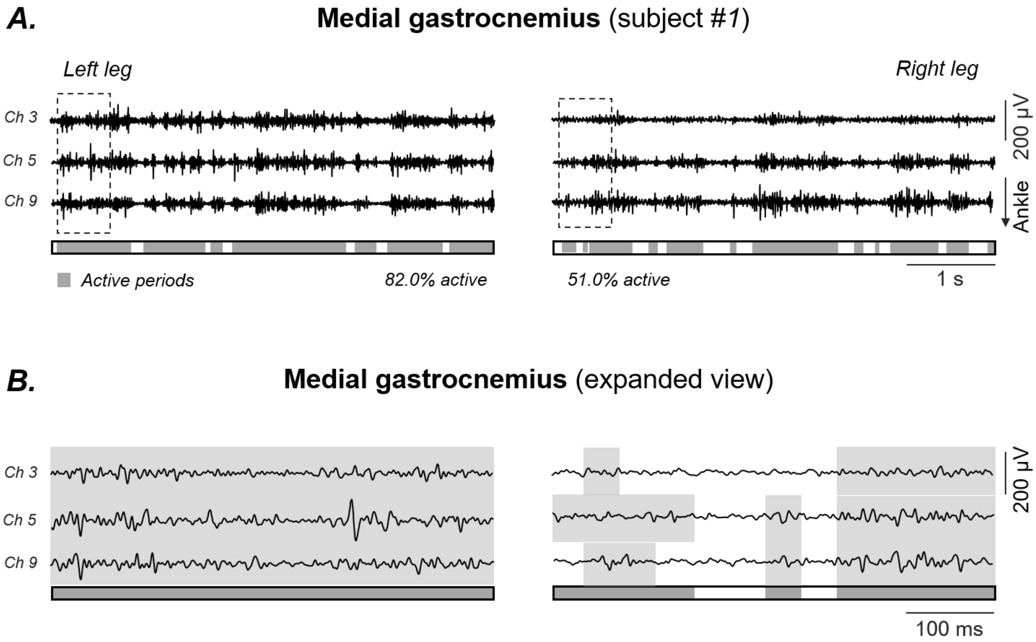

Figure 15: *A*, example of single-differential EMGs recorded by channels 3, 5 and 9 from the left and the right medial gastrocnemius of a single, representative participant. *B*, shows an expanded view (500 ms; dashed area) of the raw EMGs shown in *A*. Grey rectangles indicate the *active* periods identified separately per channel and for all channels (grey rectangles shown below EMGs; cf. Methods). Note different channels detected different action potentials and therefore provided different *active* periods for the right gastrocnemius. Percentages denote the relative number of *active* periods (i.e., duration of muscle activity) throughout the whole (60 s) standing test.

Correlation between CoP lateral position and side-differences in muscle activity

Associations between side-differences in *active* periods and CoP lateral position were muscle dependent. Subjects whose CoP was on average located closer to the right leg activated more frequently their right, gastrocnemius muscle (Fig. 17A; Pearson *R*=0.60; *P*=0.035; *N*=12 subjects). For the soleus muscle, no significant correlation between the ratio of proportion of *active* periods and CoP lateral position was observed (Fig. 17B; R= -0.10; P=0.75; N=12).

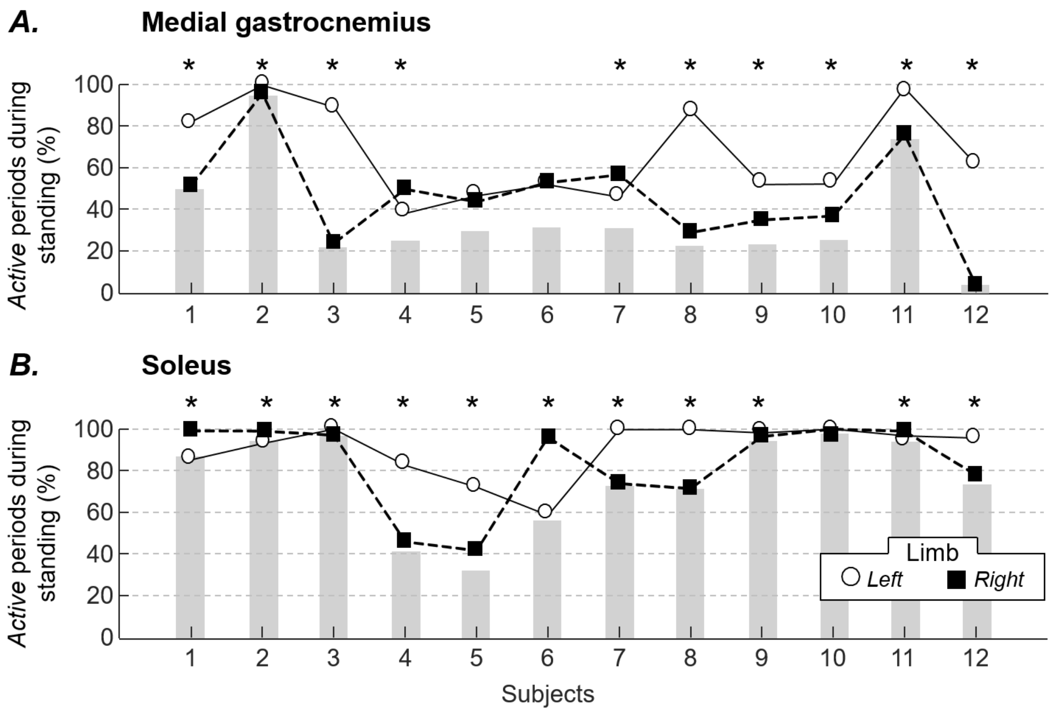

Figure 16: Asymmetries in the duration of plantar flexors' activity. The relative duration of activity of the left (circles) and right (squares) medial gastrocnemius (*A*) and soleus (*B*) muscles is shown for the 12 participants tested. Vertical, grey rectangles indicate the relative amount of the standing duration within which muscles in both legs were active concurrently. Asterisks indicate significant differences in the duration of activity between the legs (P<0.05).

5.5 Discussion

In this study we investigated whether the duration of medial gastrocnemius and soleus activity differs between legs while healthy, young subjects stood at ease. Representative estimations of the duration of activity were obtained by sampling surface EMGs from different regions within each muscle. Statistical analysis revealed that subjects activated their left and right calf muscles for different durations during standing. Side differences in the duration of muscle activity were observed for at least 10 out of the 12 participants tested, ranging from 2% to 65%

(Fig. 16). Exclusively for gastrocnemius, these side differences were positively correlated with the mean CoP lateral position (Fig. 17). Our results suggest therefore that surface EMGs may provide different estimates of the timing of plantar flexors' activity if collected unilaterally during standing.

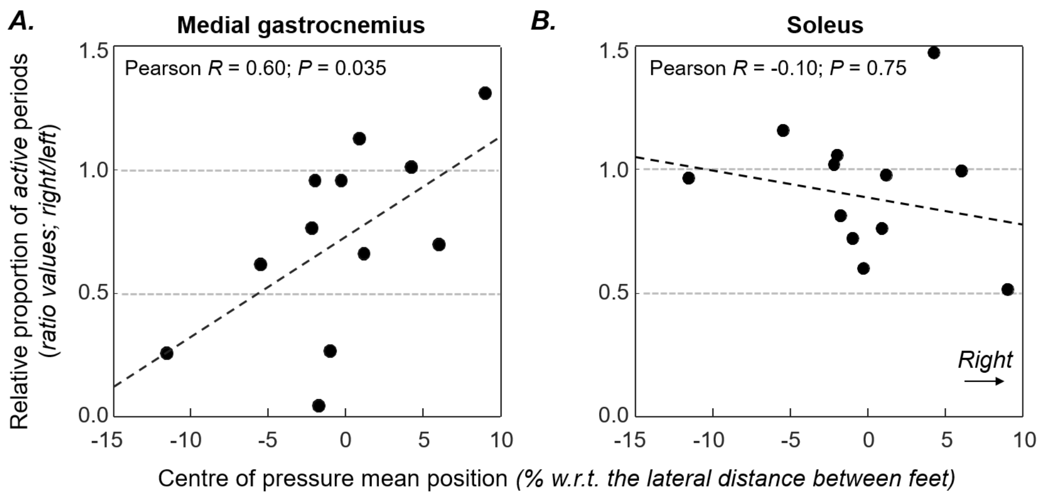

Figure 17: Side differences in the duration of muscle activity and centre of pressure position. Scatter plots are shown, with the ratio (right/left) of the duration of medial gastrocnemius (*A*) and soleus (*B*) activity plotted in the y axis and the centre of pressure (CoP) position in the frontal plane plotted in the x axis. CoP position was normalised with respect to the lateral distance between feet (cf. Fig. 1). Regression (dashed) lines were drawn for clarity.

Were plantar flexors active for similar durations between legs?

Individual results indicate gastrocnemius and soleus muscles were active for different durations between legs for 10 and 11 out of 12 subjects tested, respectively. We analysed subjects separately because there was no reason to group them according to body side. Our hypothesis that plantar flexors in both limbs would be activate for different durations during standing was based on side-differences in the amplitude of surface EMGs [139,142]. Even though subjects may alternate the distribution of body weight between limbs [145,146], in particular during prolonged standing [141], we are not aware of any evidence suggesting subjects should active consistently muscles in either right or left leg. Indeed, our results suggest some subjects activated for longer durations their right plantar flexors whereas others showed the opposite (Fig. 16). Differences in the duration of muscle activity between legs were also variable between subjects, ranging from 4% to 65% for gastrocnemius and from 2% to 40% for soleus. Regardless of these inter-individual differences in the duration of activity, asymmetries in the duration of the period through which gastrocnemius and soleus muscles were active were generally observed (Fig. 16). Results shown in Fig. 16 seem therefore to support the notion that muscles in both limbs were elicited for different durations during standing.

Side differences in the duration of activity differed between muscles. Subjects showing the greatest differences in the duration of gastrocnemius and soleus activity between legs were different (Fig. 16). Specifically concerning gastrocnemius, the duration of activity was associated with the CoP mean, lateral position; subjects standing closer to the right leg activated for longer duration their right gastrocnemius muscle and vice-versa (Fig. 17). This observation is consistent with the documented contribution of gastrocnemius muscle to the production of ankle inversion torque [111,147]. Similar reports on a systematic contribution of soleus to ankle inversion-eversion moments were not found, possibly because the soleus line of action is directed more closely to the midline of the foot than that of gastrocnemius [147]. Asymmetries in the timing of activation of gastrocnemius though not of soleus were partly explained (36%; Fig. 17) by lateral differences in the CoP mean position, which may be therefore associated with the uneven loading of both legs [148]. Corroborating this differential muscle response, Henry and colleagues [110] observed the medial gastrocnemius responds to surface translations directed over a larger, oblique range than soleus (cf. their Fig. 3). When drawing considerations on differences between muscles from current results, it should be noted however we were able to sample from a small, medial region of the large soleus muscle (Fig. 14), not concealed by gastrocnemius. As discussed by Agur et al. [149], in virtue of architectural differences within soleus, surface EMGs collected from the muscle medial region may reflect a predominant, plantar flexion action. Regardless of the actual, predominant action of the soleus region sampled in this study, asymmetries in the duration of activity were observed (Fig. 16). Factors other than CoP mean lateral position may contribute to explaining side differences in the duration of plantar flexors' activity. While the identification of these sources urges further investigation, current results suggest inferences on the timing of muscle activation during standing may not proceed from EMGs collected unilaterally.

What are the implications of asymmetric activation of plantar flexors?

Before commenting on the implications of current findings, we would like to mention we assessed side differences in the timing rather than in the degree of muscle activity. Two reasons motivated our decision. First, averaging the amplitude of EMGs across the whole standing duration would provide a biased indication on the degree of muscle activity; low, average EMG amplitude may not necessarily indicate low degree of activation but e.g. alternated periods of rest and activity [14,143]. Most importantly, the timing of muscle activity, as quantified from surface EMGs, has provided substantial contribution to our understanding of the human, postural control. It seems to discriminate well the pattern of activation of different muscles [33] and populations [14,95] during standing. It has also driven the notion that standing stabilisation may be achieved well by alternating periods of muscle silencing and activation [28,33]. Asking therefore whether surface EMGs detected from muscles in either leg reflect muscles in both legs during standing seems of general relevance.

Results presented here have direct, practical and methodological implications. On one hand, our results corroborate the general view that gastrocnemius and soleus remain active for different durations during standing; despite the inter-individual difference in the stiffness of the ankle joint [6], previous studies consistently report that gastrocnemius and soleus are activated respectively intermittently and continuously during standing [26,33]. On the other hand, our results (Fig. 16) indicate that surface EMGs detected bilaterally do not provide equal estimates of the duration of plantar flexors activity. Even the soleus muscle, which is often observed to be active continuously during standing [26,34], showed a variable duration of activity between legs for some subjects (Fig. 16*B*). These results are not necessarily in contrast with the view that humans sway as an inverted pendulum during standing. While experimental observations of CoP and centre of gravity sways have confirmed the inverted pendulum, standing model [4], the inverted pendulum assumption does not seem to justify stating ankle muscles in both legs are activated similarly (Fig. 16). More specifically, according to results presented here, the active participation of plantar flexors to the correction of bodily sways may be either under- or over-valued, depending on the body side from which surface EMGs are detected. Drawing inferences on the neural mechanisms governing the activation of plantar flexors during standing may therefore require the detection of EMGs from both legs.

Chapter 6

Is the attenuation effect on the ankle muscles activity from the EMG biofeedback generalized to – or compensated by – other lower limb muscles during standing?

6.1 Abstract

Biofeedback based on electromyograms (EMGs) is a promising technique to reduce exaggerated muscle activity. For example, it has been recently shown that healthy subjects are able to suppress the plantar flexors' activity when provided with EMG-audio feedback during standing without threatening stability. Whether however the effect of EMG-biofeedback on the ankle muscles' activity generalizes to – or is compensated by – other muscles not included for the biofeedback is still an open question we address here. Fourteen young individuals were asked to stand barefoot comfortably with eyes open. Three 60 s standing trials were applied: (i) without EMG-audio feedback, (ii) with EMG-audio feedback from medial gastrocnemius and soleus and (iii) with EMG-audio feedback from medial gastrocnemius and tibialis anterior muscles. Surface EMGs from the soleus medial aspect, medial

gastrocnemius, tibialis anterior, vastus medialis, vastus lateralis, semitendinosus, and biceps femoris of both legs were sampled with a pair of surface electrodes. The Root Mean Square (RMS) of EMGs over the whole standing duration was averaged across sides to assess the degree of postural activity for each muscle. In relation to standing at ease, our main findings revealed EMG-audio feedback from medial gastrocnemius and soleus muscles decreases the EMG level of plantar flexors (~15%; $P < 0.05$), did not change the level of tibialis anterior's activity ($P > 0.05$) and increases the amplitude of surface EMGs sampled from vastus medialis and lateralis muscles (~ 40%; $P < 0.05$). In contrast, EMG-audio feedback from medial gastrocnemius and tibialis anterior muscles leads to a reduction in the level plantar flexors' activity (~5%; $P < 0.05$) without increasing the amplitude of EMGs collected from the tibialis anterior, vastus medialis, vastus lateralis, semitendinosus, and biceps femoris ($P > 0.05$). These alterations in the level of postural activity were accompanied by a posterior shift of the mean CoP position (~5% of the foot length projection; $P < 0.05$ when using the feedback from both plantar flexors) and no significant changes in the size of CoP sways ($P > 0.05$) regardless of EMG biofeedback condition. These results indicate the attenuation effect of EMG-audio feedback on calf muscles is not compensated by other lower limbs muscles not included for the feedback depending on the biofeedback condition tested here. Therefore, the EMG-audio feedback may be a promising technique to assist individuals in reducing excessive muscle activity without overloading other lower limbs muscle during standing.

6.2 Introduction

EMG biofeedback is a promising technique to re-educate the pattern of muscle activity during a task. Prior accounts have observed individuals while receiving continuously real-time information about the own level of muscle activity are able to attenuate or extend the level of muscle activity. For example, EMG biofeedback has been effective to directly assist patients with chronic muscle pain in reducing the excessive level of muscle activity associated likely with the muscle pain (e.g., in the neck-shoulder; [25]). Likewise, reductions in the level of wrist flexor muscle spasticity at the hemiplegic arm in patients following stroke were also observed after an EMG biofeedback training [81]. Alternatively, EMG biofeedback may also assist individuals in increasing the degree of muscle activity [80,91]. Collectively, these results suggest individuals could learn to change their pattern of muscle activity with the EMG biofeedback-based training; potential technique to re-educate muscle activity.

Recently, EMG biofeedback has been proposed for reducing the exaggerated muscle activity while individuals stand at ease. To author knowledge, Vieira et al. [72] investigated for the first time the influence of EMG-audio feedback protocol on the ankle muscles' activity during standing. In relation to standing at ease, the authors observed young subjects were able to decrease by 5% and 10% the degree of medial gastrocnemius and soleus muscles' activity, respectively, with the EMG-audio feedback. This attenuation effect on the calf muscles' activity from the EMG-audio feedback did not lead to excessive postural sways and greater level of antagonistic activation. These results indicate EMG-audio feedback could assist young subjects in more efficiently controlling leg muscle activity without hindering postural stability. As recently put forward in the literature, reduction of unnecessary muscle activation may provide possible advantages for the control of posture, such as: (i) proprioceptive feedback without noise from muscle contraction [30]; (ii) body stability from an economical, fatigue-resistant control [36]; (iii) decreases in postural rigidity [56]. Thus, the possibility of minimizing muscular effort without excessively increasing the size of postural sway may posit an innovative training paradigm specifically targeting the fine control of postural sways, with particular relevance for the training of populations who activate exaggeratedly their muscles during standing, e.g. aged individuals.

Additional investigations, however, might be relevant to understand the effects of this technique on the level of activity of lower limbs muscles. Researchers have long been indicated the assessment of response generalization of EMG biofeedback should be considered in the repertoire of EMG biofeedback protocols aimed at reducing the general level of muscle activity [87]. The main concern is whether the attenuation effect on an single muscle from EMG biofeedback is manifested at – or compensated by - other muscles during the task [89]. Prior accounts have demonstrated that an attenuation effect and/or compensation effect from EMG biofeedback protocols may occur between muscles depending on the motor task. On one hand, EMG feedback from a targeted muscle seems sufficient to transfer the relaxation effect to the contralateral muscle during computer work conditions [24,93]. On the other hand, limitation to transfer the relaxation effect to not-synergistic muscles may occur when biofeedback is provided from a single muscle during computer work conditions [93]. Moreover, other muscles not included in the feedback may compensate the attenuation effect on the targeted muscle from the EMG biofeedback in strictly controlled biomechanics conditions (i.e., isometric shoulder abductions [94]). Collectively, these results suggest it is relevant to assess the activity of other muscles not included in the biofeedback to prove the effectiveness of innovative biofeedback protocols in reducing the level of muscle activity, e.g. during standing.

This study aimed at investigating therefore whether the attenuation effect on calf muscles from the EMG-audio biofeedback is generalized to – or compensated by - other lower limbs muscles during standing. If the attenuation effect of EMG-audio feedback on calf muscle' activity is not compensated by other muscles, we expect to observe: (i) bipolar EMGs collected from the ankle muscles with lower amplitude than those sampled during standing at ease; and (ii) bipolar EMGs collected from the thigh muscles with amplitude values within standing at ease levels, since thigh muscles seem to contribute marginally for the control of standing balance in healthy, young subjects [11,14].

6.3 Methods

6.3.1 Participants

Fourteen young individuals (12 men; range values: age 21–36 years; body mass 58–88 kg; height 162–190 cm) were tested and provided written informed consent before participating in the study. The experimental procedures considered in this study conformed to the latest Declaration of Helsinki and were approved by the Regional Ethics Committee (Commissione di Vigilanza, Servizio Sanitario Nazionale—Regione Piemonte—ASL1—Torino, Italy). Volunteers did not report any balance impairments, neurological disorders, muscular injuries, or the intake of medications that could affect their standing balance at the occasion of experiments.

6.3.2 Experimental protocol

Participants were instructed to stand upright barefoot on a force plate, with eyes open and arms alongside the body. They positioned their feet at a comfortable orientation and distance, while keeping heels and toes at the same position along the force plate anterior-posterior axis. Prior to starting experiments, the contour of both feet was marked on the force plate to ensure participants would keep the same feet position throughout the standing tests.

Three standing trials were applied. In the first task, subjects were asked to sway as much as possible back and forward for 40s. They were specifically instructed to sway their body only from ankle movement, without lifting up their fore and rear foot and at their preferred speed. This postural task was chosen to ensure a somewhat high degree of ankle and thigh muscles' activity during standing. Then, this postural task was considered for the normalization of surface EMGs, and for providing EMG-audio feedback to the participants (see subsection 6.3.5). In the second task, referred to as standing at ease, subjects were engaged in active conversation for 60s to ensure they would not concentrate on their posture while standing and thus avoid any voluntary change in muscle activity [102]. In the third task, participants were instructed to reduce the volume of an audio signal proportional to the level of activity of their ankle muscles without changing standing posture. EMG-audio feedback based on the level of activity of the right and left (i) medial gastrocnemius and soleus muscles and (ii) medial gastrocnemius and tibialis anterior muscles were tested. We decided to provide EMG-audio feedback to the subjects related to their right and left calf muscles to avoid a biased feedback of muscle activity, since side-differences in the calf muscles' activity were observed previously while individuals stood at ease (see Chapter 5). The motivations for testing the effects of an audio feedback based on EMGs collected from medial gastrocnemius and tibialis anterior muscles on the lower limbs muscles' activity were: (i) potential impact on individuals with balance impairments (e.g., aged subjects) who may activate to a greater extent and for a prolonged duration mainly medial gastrocnemius and tibialis anterior muscles during standing [143]; (ii) in preliminary experiments on eight subjects who participated in the current study, sporadic modulations in the bipolar EMGs collected from tibialis anterior muscles were observed during standing. Before each EMG-audio feedback condition, a brief period of familiarization with the audio stimulus was given to the participants. For all standing tasks, subjects were not allowed to move their head, trunk and upper and lower limbs. Trials started over in the case any gross movements was perceived by the evaluator. Two minute intervals were applied between standing trials, which were selected at random order. The standing at ease task and each EMG-audio feedback condition were applied two times.

6.3.3 Electrodes' positioning

A single pair of adhesive surface electrodes (24 mm diameter, Spes Medica, Battipaglia, Italy) was positioned on ankle and thigh muscles of both legs (Figure

18). Inter-electrode distance was defined according to the total tissue (subcutaneous + muscle) thickness for each muscle tested using ultrasound imaging as described in the following. We expected such inter-electrode distance provides more representative information of muscle activity with negligible activity from nearby muscles (i.e., crosstalk) [21]. At the ankle level, medial gastrocnemius, soleus (medial aspect) and tibialis anterior muscles were studied. In the medial gastrocnemius, the center of distal electrode was positioned medially to the most distal part of junction between gastrocnemius' heads and, the bipolar system was positioned parallel to the line connecting the junction between medial gastrocnemius and Achilles tendon and the popliteal fossa. The positioning of a pair of surface electrodes moderately spaced on the medial gastrocnemius' distal-medial region may sample bipolar EMG more representative of muscle activity, since active postural units may reside more distally in the medial gastrocnemius [20]. In the medial aspect of soleus muscle, the distal electrode was positioned 3 cm distally from the junction between the Achilles tendon and the medial gastrocnemius [23] and the center of the proximal electrode was positioned at an angle of 45 degrees to the line connecting the junction between gastrocnemius' heads and the calcaneus tip. The bipolar system on the tibialis anterior muscle was positioned parallel to the muscle longitudinal axis and located 1 cm laterally to the tibial crest; the center of the proximal electrode was positioned 2 cm distally to the head of fibula bone. As propagating potentials may be observed in surface EMGs collected from multiple surface electrodes positioned serially on the distal-medial region of tibialis anterior muscle [143], we expected that a pair of surface electrodes moderately spaced and positioned on the medial-proximal region of tibialis anterior provides bipolar EMG representative of different muscle fibers. For each thigh muscle, the pair of surface electrodes was positioned approximately on the most prominent bulge of the muscle belly [11,14,130] and with an inter-electrode distance equivalent to total tissue thickness. Given the pennate architecture of thigh muscles[18], such positioning is expected to provide bipolar EMGs more representative of muscle activity.

Ultrasound imaging was used to estimate total tissue (subcutaneous + muscle) thickness as well as to identify gastrocnemii junction and their myotendinous junction (cf. Supplementary Material in Vieira et al., 2010b). Briefly, parasagittal images of ankle and thigh muscles were taken with a linear, ultrasound probe (10MHz, 4cm length; Echo Blaster 128, Telemed Ltd., Vilnius, Lithuania) located approximately on the same electrodes' positioning, with participants lying comfortably on a padded bed. Total tissue thickness was estimated as the distance between the skin/fat layer over the muscle and the muscle deep aponeurosis Thickness measurements were taken from the central region of the ultrasound images. Pairs of surface electrodes were positioned on each muscle after cleaning the skin with abrasive paste.

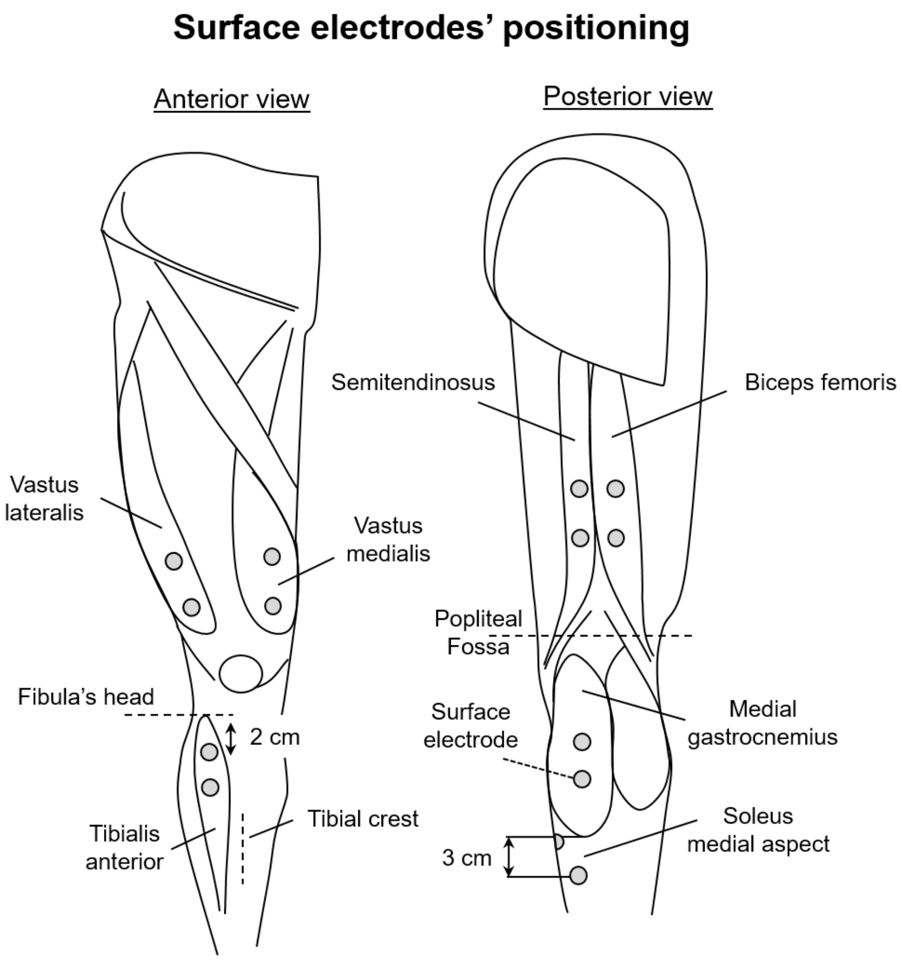

Figure 18: Electrodes' positioning. Schematic illustration showing the positioning of pair of electrodes for the tibialis anterior, vastus lateralis and vastus medialis (left), medial gastrocnemius, soleus medial aspect, semitendinosus and biceps femoris (right) of the right leg.

6.3.4 Electromyographic and stabilometric recordings

Bipolar EMGs were recorded with a wearable, wireless system (180 V/V gain; 10–500 Hz bandwidth amplifier; DuePro system, OTBioelettronica and LISiN, Politecnico di Torino, Turin, Italy). EMGs were digitized at 2,048 Hz with a 16 bits A/D converter. Center of pressure (CoP) coordinates in the sagittal and frontal planes were computed from the ground reaction forces supplied by a piezoelectric force-plate (9286AA Kistler, Zurich, Switzerland). Reaction forces were sampled at 2,408 Hz using a multifunction data acquisition device (16 bits A/D converter; USB-6210, National Instruments, Austin, USA). A custom Matlab script (MathWorks Inc., MA, USA) was considered to sample synchronously bipolar EMGs and reaction forces.

6.3.5 Modulation of audio signal from bipolar, surface EMG

The audio stimulus (sinusoidal signal) was modulated in amplitude and frequency on the basis of EMG amplitude with the same procedure proposed by Vieira et al. [72] to provide subjects with EMG-audio feedback of surface EMGs collected from calf muscles with linear arrays of surface electrodes (cf. Figure 1 in [72]).

Depending on the biofeedback condition, bipolar surface EMGs collected from both medial gastrocnemius and soleus muscles or from both medial gastrocnemius and tibialis anterior muscles were considered. Bipolar surface EMGs from the selected muscles (Figure 19A) were: (i) full-wave rectified and smoothed with a low-pass, 4th order Butterworth filter (3Hz cut-off frequency) obtaining the EMG envelopes; (ii) each EMG envelope was normalized with respect to EMG envelope identified during the reference condition (voluntary sways; see 6.3.2). The 5th and the 95th percentiles of the reference EMG envelope were respectively used for the subtraction of the background noise and for the amplitude normalization of the EMG envelopes; (iii) the normalized EMG envelopes were averaged to provide an EMG envelope representing the activity of muscles (Figure 19B); (iv) the averaged envelope was resampled at 8000 samples/s. The amplitude of the audio stimulus was modulated from a sigmoid function, while the frequency of this signal was modulated from a linear function as detailed below (Figure 19B). Previous evidence has shown from this coding, EMG-audio feedback of surface EMGs collected from calf muscles assists individuals in reducing the muscle effort at ankle joint during standing [72].

The amplitude a[n] of the audio signal was modulated as:

$$a[n] = \frac{1}{\left(1 + e^{-c(EMG[n]-b)}\right)} \quad (1)$$

$$b = \frac{T_h}{2} \quad (2)$$

$$c = \frac{4.6}{b} \quad (3)$$

where EMG[n] corresponds to the samples of averaged EMG envelope, after background noise subtraction and normalization, and T_h stands for the threshold defining the sensitivity of the EMG audio modulation. For example, when the EMG envelope reaches T_h the intensity of the audio stimulus reaches 99% of its maximal value. Once a very sensitive audio stimulus likely makes the task of reducing the level of muscle activity from EMG-audio feedback very hard (i.e., audio volume and frequency easily saturate [72]), EMG-biofeedback sensitivity was defined by setting T_h to 90% of the 95th percentile of the distribution of the averaged EMG envelope obtained during the voluntary sways. By comparing audio stimulus with different sensitivities while individuals stood with EMG-audio feedback, prior evidence reported such sensitivity (T_h to 90% of voluntary sways) was not accompanied by significant posterior shifts of postural sway and increased antagonist activity in relation to standing at ease [72].

The frequency $f[n]$ of the sinusoidal, audio signal was modulated according to:

$$f[n] = \begin{cases} EMG[n]m + 100, & EMG[n] < T_h \\ 450, & EMG[n] \geq T_h \end{cases} \quad (4)$$

where the angular coefficient m being defined as:

$$m = \frac{350}{T_h} \quad (5)$$

EMG biofeedback was provided to participants in real time. A custom Matlab script (MathWorks Inc., MA, USA) was written to output the modulated, audio signal into the soundcard of a personal computer. The audio stimulus was then transmitted wirelessly to headphones worn by participants. The time taken to compute and average envelopes and modulate the sinusoid ranged from 10 to 20ms.

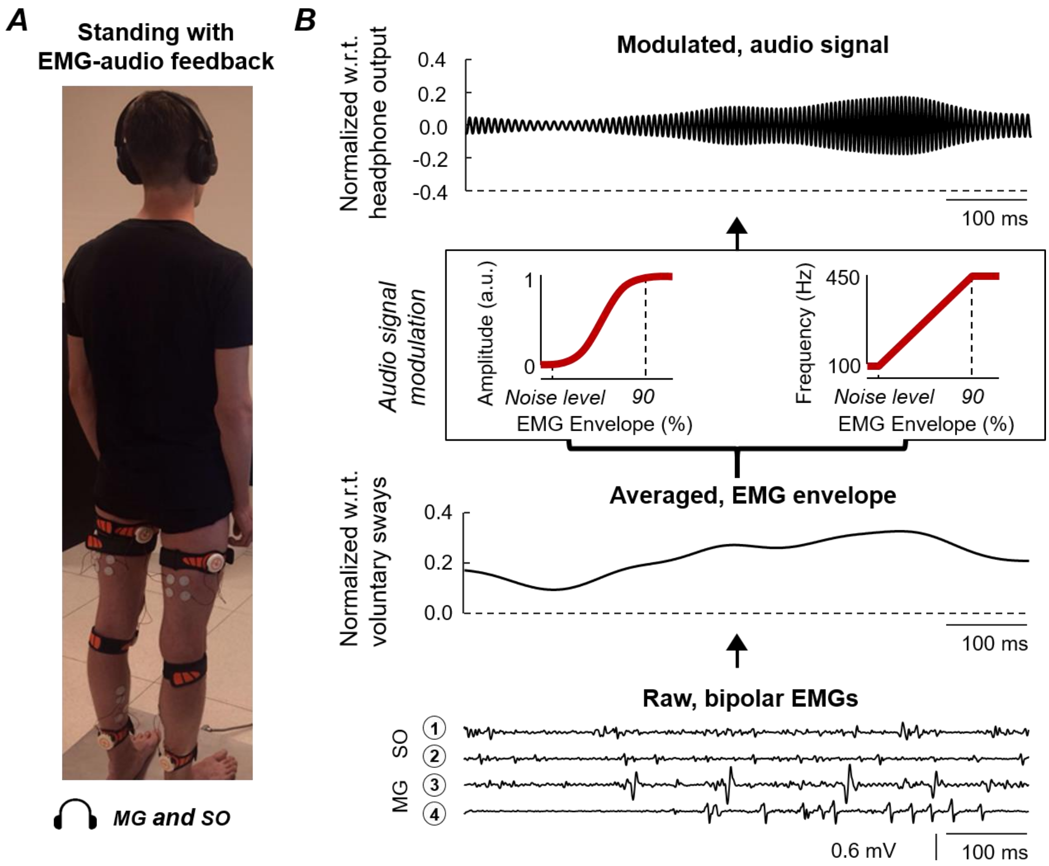

Figure 19: (*A*) Standing with EMG-audio feedback, where participants were asked to reduce the volume of an audio signal proportional to the level of activity of their ankle muscles without changing standing posture. (*B*) Modulating audio signal from bipolar EMGs. *B* shows a short epoch (700 ms) of raw bipolar EMGs collected from both soleus (SO, EMG signals 1 and 2) and medial gastrocnemius (MG; EMG signals 3 and 4) while participant stood at ease and exhibits the averaged, EMG envelope computed from these bipolar EMGs, which its amplitude was considered for modulating the audio stimulus (sinusoidal signal). The amplitude and the frequency of this audio signal were respectively modulated according to a sigmoid and linear function. Note the correspondence between the amplitude of bipolar EMGs, the EMG envelope and the modulated audio signal.

6.3.6 Quantifying variations in the degree of muscle activity

The Root Mean Square (RMS) was used to quantify the amplitude of surface EMGs collected from the ankle and thigh muscles during the standing conditions. Bipolar EMGs were band-pass filtered with a fourth order Butterworth filter (20–350Hz cutoff; zero lag, bidirectional filter). Then, the RMS amplitude of EMGs was computed over the whole standing duration (60s) and it was averaged across sides to assess the degree of postural activity for each muscle. Averaging the amplitude of EMGs across the whole standing duration would provide an overall indication of any increase in the degree of muscle activity throughout standing condition. The RMS amplitude of EMGs was averaged across the standing trials.

6.3.7 Quantifying variations in the center of pressure

Global posturographic parameters of CoP displacement were considered to assess whether subjects change their postural sway due to EMG-audio feedback conditions. Specifically, alterations in the size of CoP sways and in the mean CoP position (during the whole standing duration) were quantified in the sagittal plane. Our interest in evaluating CoP variations in the anterior-posterior direction was due to previous study demonstrating the EMG biofeedback affects the CoP position and displacements in the sagittal rather than in the frontal plane while young subjects stood at ease [72]. The standard deviation was computed over 2 s epochs to evaluate the effect of EMG-audio feedback on the size of individual CoP sways [72]. The mean CoP position in the anterior-posterior direction was normalized with respect to the foot length projection for assessing how much the mean CoP position travels in relation to the boundary of foot (i.e., support base) between standing conditions. The size of the support base in the anterior-posterior axis was defined as the distance between the tip of the third metatarsal head and the tip of the calcaneus bone projected in the anterior-posterior direction (cf. Figure 1 in [143]). The standard deviation and the mean CoP position were averaged across the standing tests. CoP data was 50 Hz low-pass filtered with a second order Butterworth filter to attenuate high-frequency noise.

6.3.8 Statistical Analysis

After verifying that the data distribution for the RMS amplitude of bipolar EMGs computed from some muscles was not Gaussian (Shapiro-Wilk's W-test, $P < 0.05$ in nine cases), non-parametric statistics were used to compare the degree of muscle activity between standing conditions. The effect of EMG-biofeedback on the degree of muscle activity was assessed with Friedman ANOVA by ranks, with standing conditions as repeated measures. Paired comparisons were made with Wilcoxon signed-rank test [144], considering the significance level of 5 %. The same statistics was applied to compare the CoP mean position and the size of CoP sways (i.e., standard deviation) between standing conditions in the sagittal plane.

6.4 Results

The level of lower limbs muscles' activity with EMG-audio feedback

Bipolar EMGs collected from the ankle and thigh muscles of a representative participant were clearly different between standing at ease and EMG-audio feedback conditions (Figure 20). For the ankle muscles, Figure 20 shows bipolar EMGs with greater amplitude for the plantar flexors during the standing at ease with respect to EMG-audio feedback conditions for a representative participant. For the tibialis anterior, however, EMGs with low amplitude were observed between standing conditions. Contrary to the ankle muscles, greater EMGs were observed for the thigh muscles when EMG-audio feedback was provided from both medial gastrocnemius and soleus muscles in relation to the other conditions. More specifically, EMGs with markedly high amplitude were observed for the vastus medialis and vastus lateralis while standing with EMG-audio feedback from both plantar flexors. High modulations in the EMG amplitude for the semitendinosus and biceps femoris muscles, however, were not often observed between standing conditions for this individual.

Group data revealed that the RMS amplitude of bipolar EMGs, i.e. level of muscle activity, varied significantly with biofeedback conditions during standing (Friedman ANOVA main effect; $P < 0.05$ for soleus, medial gastrocnemius, vastus medialis and vastus lateralis). For soleus, RMS amplitude decreased of about 15% when EMG-audio feedback was provided from plantar flexors (median, interquartile range; 31.33, 25.93 - 56.77%) and of about 5% with the EMG-audio feedback from plantar and dorsal flexors (medial gastrocnemius and tibialis anterior; 40.47, 32.47 - 70.32%) in relation to standing at ease (44.56, 40.13 - 68.25%, $P < 0.05$ in all cases; Figure 21). When considering the difference in the normalized RMS amplitude between standing conditions, lower RMS values were observed for fourteen (range: 1.27 - 41.90%) and ten (range: 0.82 – 27.58%) out of fourteen subjects during EMG-audio biofeedback from medial gastrocnemius-soleus and from medial gastrocnemius-tibialis anterior respectively with respect to standing at ease. For the medial gastrocnemius, EMG-audio feedback from plantar flexors reduced significantly RMS amplitude in ~10% (10.10, 8.01 - 18.56%), while EMG-audio feedback from medial gastrocnemius-tibialis anterior attenuated significantly it in ~5% (13.52, 10.07 - 19.40%) when compared with standing naturally (18.30, 12.11 - 25.67%, $P < 0.05$ in all cases; Figure 21). While thirteen participants reduced the level of medial gastrocnemius' activity with the EMG-audio biofeedback from medial gastrocnemius-soleus (range: 0.26 - 20.08%), ten individuals activated to a lesser extent their medial gastrocnemius during the EMG-audio biofeedback from medial gastrocnemius-tibialis anterior (range: 0.32 - 17.46%) in relation to standing at ease. Moreover, between biofeedback conditions, lower RMS amplitude was observed for soleus (~10%) and medial gastrocnemius (~3.5%) with EMG-audio feedback from plantar flexors than with EMG-audio feedback from plantar and dorsal flexors for at least ten participants ($P < 0.03$ in all cases). For the tibialis anterior, no statistical differences in the RMS amplitude were revealed between standing conditions (Friedman ANOVA, $P = 0.11$; Figure 21).

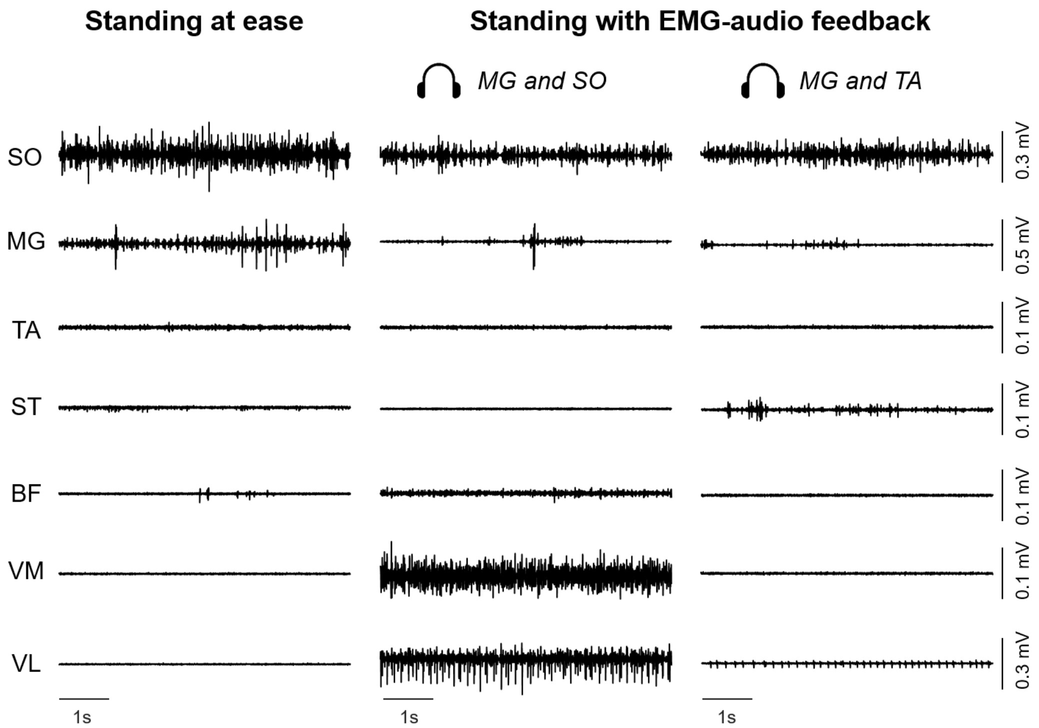

Figure 20: Modulations in the ankle and thigh muscles' activity with EMG-audio feedback. Bipolar EMGs collected from the soleus (SO) medial aspect, the medial gastrocnemius (MG), the tibialis anterior (TA), the semitendinosus (ST), the biceps femoris (BF), the vastus medialis (VM) and vastus lateralis (VL) of a representative participant are shown during standing at ease and standing with EMG-audio feedback. Signals are shown over a relatively short epoch (5s) to facilitate observing the modulations in bipolar EMG amplitude for the different standing conditions.

Different from ankle muscles, significantly higher RMS values (~40%) were revealed for vastus medialis (50.45, 18.97 - 65.07%) and vastus lateralis (47.52, 26.24 - 69.63%) while subjects stood with EMG-biofeedback from plantar flexors than standing naturally (vastus medialis: 12.25, 3.74 - 40.48%; vastus lateralis: 8.75, 4.79 - 39.41%; $P < 0.01$ in all cases; Figure 21). Notwithstanding markedly high RMS values during EMG-audio feedback from plantar flexors, there was no significant difference in the RMS amplitude for vastus muscles between EMG-audio feedback when provided from medial gastrocnemius and tibialis anterior muscles and standing at ease ($P > 0.06$ in all cases). Finally, no significantly differences were found in the RMS amplitude of bipolar EMG collected from semitendinosus and biceps femoris between standing conditions (Friedman ANOVA, $P > 0.29$ in both cases).

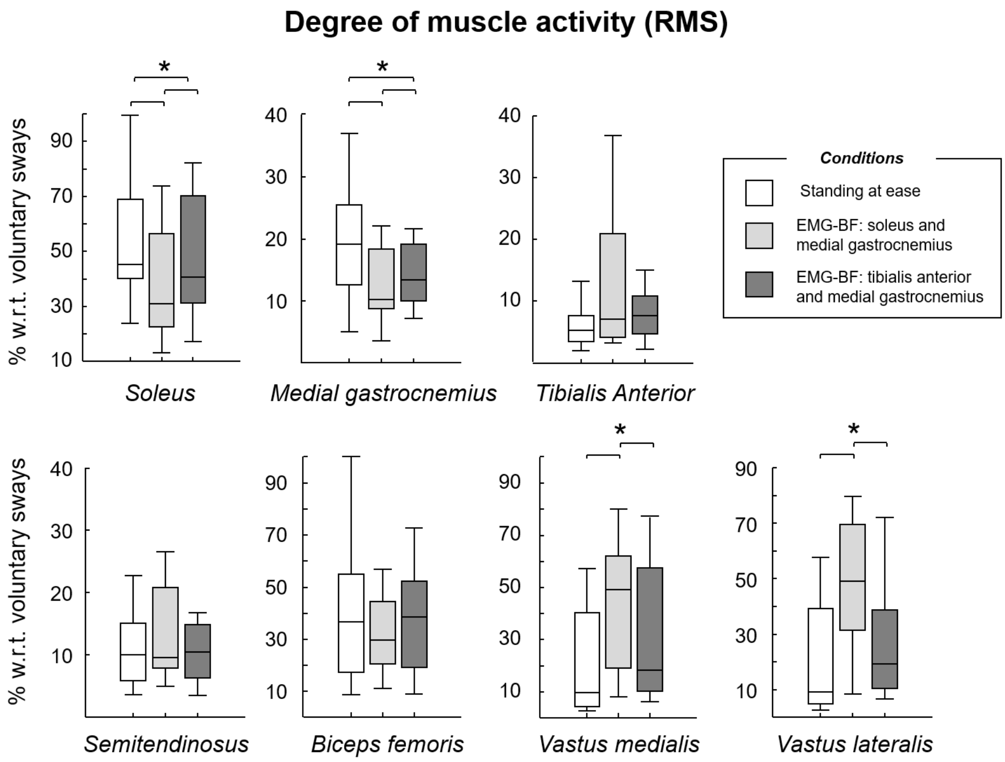

Figure 21: Differences in the degree of ankle and thigh muscles' activity during EMG-audio feedback conditions. Boxplots are shown for the RMS amplitude of surface EMGs collected from the ankle and thigh muscles during standing conditions. Whiskers denote non-outlier range and asterisks indicate significant differences between conditions ($P < 0.05$).

Alterations in the CoP displacement with EMG-audio feedback

When considering data from all participants for the posturographic parameters of CoP displacement, significant differences were revealed for the mean CoP position between standing conditions in the sagittal plane (Friedman ANOVA main effect; $P < 0.01$). Wilcoxon signed-rank test showed that the mean CoP position normalized by foot length projection was localized closer to the heel while individuals stood with the EMG-audio feedback from plantar flexors (median, interquartile range; 37.27, 34.67 – 40.62%) in relation to that estimated during standing without (42.00, 38.32 – 44.49%) and with the EMG-audio feedback from plantar and dorsal flexors (38.83, 35.46 – 49.10%; $P < 0.01$ in all cases). Contrary to the mean CoP position, the size of CoP sways did not change (Friedman ANOVA main effect; $P = 0.29$) between standing at ease (1.97, 1.63 – 3.05 mm), with EMG-audio feedback from plantar flexors (2.41, 1.90 – 2.83 mm) and with EMG-audio feedback from plantar and dorsal flexors (1.96, 1.68 – 2.23 mm).

6.5 Discussion

In this study, we specifically asked whether the attenuation effect of EMG-audio biofeedback on the level of calf muscles' activity is generalized to – or compensated by - other lower limb muscles during standing. Surface EMGs were collected from different ankle and thigh muscles of both legs with pairs of surface electrodes to estimate the degree of muscle activity. We hypothesized EMGs collected from the calf muscles with lower amplitude during biofeedback conditions than standing at ease, and those sampled from the thigh muscles with amplitude values within standing at ease levels, regardless of biofeedback condition. The main findings of this study were: (i) the RMS amplitude of bipolar EMGs collected from medial gastrocnemius and soleus muscles decreased significantly during the EMG biofeedback conditions tested here in relation to standing at ease; (ii) the RMS amplitude of EMGs sampled from the thigh muscles did not change during the EMG-audio feedback from medial gastrocnemius and tibialis anterior muscles when compared with standing naturally. These results suggest with the feedback from the medial gastrocnemius and tibialis anterior muscles, EMG-audio feedback assists young individuals in reducing the level of ankle muscles' activity without increasing the level of activity in other lower limb muscles.

Methodological issues on surface EMG detection for EMG biofeedback

In the current study, we provided EMG-audio feedback based on calf muscles' activity from both lower limbs and sampled surface EMGs with a pair of surface electrodes. Firstly, the motivation to provide EMG-audio feedback involving surface EMGs recorded from ankle muscles of both legs was based on our previous study (see Chapter 5 in this Thesis) showing side-differences in the ankle muscles' activity while young subjects stand at ease. Depending on the side where bipolar EMGs are collected from (ankle muscles of right or left leg), therefore, a biased view of ankle muscles' activity can be estimated and provided to the subjects as EMG biofeedback (cf. subjects #3 and #12 in Figure 16). On the recent view that a pair of surface electrodes may sample bipolar EMG more representative of muscle activity, at least in postural muscles during standing [21], we opted by providing EMG-audio feedback from a parsimonious electrodes system rather than the use of a high number of electrodes in order to make the task of acquiring surface EMG data from lower limbs muscles unobtrusive and less labor intensive; extremely recommended in applications of EMG biofeedback [78]. More specifically, by triggering and averaging surface EMGs collected from medial gastrocnemius with firing instants of soleus' motor units identified from intramuscular EMGs during standing, Vieira et al. [21] observed medial gastrocnemius' EMGs with high amplitude (i.e., comprising a high number of pennate fibers) and with negligible crosstalk of soleus muscle using inter-electrode distances up to 5-7 cm. Increasing the distance between electrodes may therefore be a simple alternative to surface electrodes' array for the sampling of more representative bipolar EMG from postural muscles during standing. We opted for inter-electrode distances proportional to the total tissue (muscle + subcutaneous) thickness to contain the contribution of any crosstalk marginal in the bipolar EMG collected from the targeted muscle. Collectively, through these methodological procedures, we believe to provide a more representative EMG-audio feedback information about the ankle muscles' activity to the subject and from a parsimonious electrodes system. As discussed below, new insights have been gained into the effects of EMG biofeedback on the level of activity in lower limbs muscles using this EMG-audio feedback approach.

Does EMG-audio feedback from the ankle muscles affect the degree of activity of lower limb muscles?

The level of ankle muscles' activity was reduced with the EMG-audio feedback. Individuals activated to a smaller extent their medial gastrocnemius (~5%) and soleus (~10%) muscles during both EMG biofeedback conditions in relation to standing at ease (Figures 20 and 21). From the inspection of Figure 20, for example, EMGs with lower amplitude were sampled from the medial gastrocnemius and soleus muscles of a representative individual during both EMG biofeedback conditions with respect to standing naturally. Differently, the tibialis anterior muscle was recruited on average to a similar extent between standing conditions (~5%; Figure 21). It is probably explained by the fact that young subjects rarely activate their tibialis anterior muscles during standing [14,143]. This result suggests the participants tested here did not take advantage of EMG-audio feedback to reduce the level of tibialis anterior muscle's activity. Collectively, these current findings corroborate previous results on the degree of calf muscles' activity while young subjects stood with the EMG-audio feedback from the right medial gastrocnemius and soleus muscles [72]. Vieira et al. [72] reported EMG-audio feedback resulted in a significant decrease in EMG level of plantar flexors with a marginal increase in tibialis anterior's activity compared with standing at ease. Extending the results obtained by Vieira et al. [72], our results suggest EMG-audio feedback was effective to reduce the level of plantar flexors' activity during standing when provided from soleus and medial gastrocnemius muscles or from medial gastrocnemius and tibialis anterior muscles of both legs.

Besides evaluating the effects of EMG biofeedback conditions proposed here on the activation of ankle muscles, the present study assessed the influence of EMG biofeedback on the level of thigh muscles' activity. Depending on the biofeedback condition, our results revealed increases in the level of thigh muscles' activity with respect to standing at ease. Higher RMS amplitude (~40%) was found for quadriceps muscle (i.e., vastus medialis and lateralis) with the EMG-audio feedback based on medial gastrocnemius and soleus muscles than standing at ease (see Figures 20 and 21). Such an increase in the level of vastus muscles' activity may be due, in part, to the posterior shift of body during this EMG-audio feedback condition. We observed the mean CoP position was localized closer to the ankle joint (i.e., more posteriorly ~5%) during this biofeedback condition in relation to that estimated during standing naturally. Once the body center of mass is projected vertically and slightly in front of ankle joint, moving the body center of mass posteriorly may be a potential mechanism to reduce the level of plantar flexors' activity during standing when using the EMG-audio feedback [72]. By activating the quadriceps, individuals may be attempting to prevent the body's center of mass from moving further backward. On the other hand, this EMG-audio feedback condition could lead to knee hyperextension, shifting the mean CoP position posteriorly. The greater activation of vastus muscles and non-significant changes in the level of ankle flexors in the EMG-audio feedback from medial gastrocnemius-soleus in relation to standing at ease (Figure 21) might further suggest individuals move the body center of mass more backward using more frequently the hip than ankle joint. Once the moment of inertia is smaller at hip than ankle level during standing, hip movements would produce a faster response with respect to ankle movements to attenuate the sound level from the feedback of the medial gastrocnemius and soleus' activity. However, in absence of direct measurement of the joint angles, such interpretations require further investigations. Regardless of potential mechanisms account for increases in the level of vastus muscles' activity, these results indicate the attenuation effect on the level of ankle muscles' activity obtained from this EMG-audio feedback condition was compensated by other muscles not included for biofeedback.

Different from the EMG-audio feedback from both plantar flexors, individuals did not activate to a greater extent their vastus muscles and the other thigh muscles tested here when using the EMG-audio feedback based on medial gastrocnemius and tibialis anterior muscles (see Figures 20 and 21). In seven out of 14 subjects, more specifically, changes in the RMS amplitude did not differ more than 5% for the vastus muscles between this EMG-audio feedback condition and standing naturally. The representative participant in Figure 20, for example, activated its vastus medialis at 19.70% and its vastus lateralis at 16.55% of normalization task during standing at ease, while hearing muscle activity mainly from both medial gastrocnemius, he activated its vastus medialis at 18.32% and its vastus lateralis at 14.92% of the normalization task. Sporadic efforts to prevent the body's centre of mass from moving further backward likely explain the similar degree of vastus muscles' activity between this EMG biofeedback condition and standing at ease. The non-significant, posterior shift of the CoP position might account for the reduction in the degree of plantar flexor's activity without affecting the level of thigh muscles' activity. We observed participants succeeded in reducing the level of plantar flexors' activity during this EMG-audio feedback condition with the CoP mean position located more posteriorly (~3%) than standing naturally. The smaller posterior shift of CoP mean position during the EMG-audio feedback from the medial gastrocnemius-tibialis anterior may explain the smaller differences in the level of medial gastrocnemius and soleus muscles in this biofeedback condition with respect to the feedback from medial gastrocnemius-soleus (Figure 21). The posterior shift of CoP mean position with a concomitant reduction in the level of plantar flexors' activity fits well with the notion the plantar flexors' activity tends to be attenuated when individuals sway closer to ankle joint during standing [27,33,72]. Collectively, these results indicate the EMG-audio feedback from the medial gastrocnemius and tibialis anterior muscles was effective enough to reduce significantly the level of plantar flexors' activity (cf. Figure 21) without increasing the level of activity in the other lower limbs muscles not included for the biofeedback.

Methodological implications and additional considerations

The results obtained in this study have methodological implications for applied and clinical research. We observed different effects may emerge between lower limbs muscles depending on the muscles included for the EMG biofeedback. A compensation effect on the knee muscles from the EMG-audio feedback of both calf muscles may occur, while an attenuation effect on the ankle muscles from the EMG-audio feedback based on medial gastrocnemius and tibialis anterior muscles may prevail during standing. Then, our results seem to discourage the use of EMG-audio feedback based on medial gastrocnemius and soleus muscles of both legs, if the goal is to assist individuals in more efficiently controlling leg muscle activity during standing. For example, it is well documented aged individuals activate their ankle and thigh muscles to a greater extent [12,13,143] and for a prolonged duration [14,143] with respect to young subjects during standing. From the EMG-audio feedback based on the right and left medial gastrocnemius and soleus muscles, therefore, aged subjects could activate to a greater extent their thigh muscles while decreasing the level of calf muscles' activity. By acknowledging the possible benefits related to the reduction of excessive muscle activation for the control of posture, such as (i) correction of postural sway without muscle noise [30], (ii) reduction of postural rigidity [56], (iii) concession of economical, fatigue-resistant control [36], EMG audio-feedback based on both medial gastrocnemius and tibialis anterior muscles may be more appropriate to assist individuals in reducing the unnecessary muscle activation during standing. Whether the short-term, attenuation effect of EMG-audio feedback based on the medial gastrocnemius-tibialis anterior activity in young individuals observed here is generalized to other populations (e.g., the elderly) and retained after training is the topic of future research.

Additional investigations are also necessary, however, to verify whether the compensation effect of EMG-audio feedback from both plantar flexors on thigh muscles ceases after training, prevailing the attenuation effect on ankle muscles. A progressive decrease in leg muscle activity has been observed in literature with task practice, as suggested by the shift from co-activation to reciprocal activation of ankle muscles after repeated and predictable perturbations of stance [134]. If this assumption holds, the training with EMG-audio feedback from both medial gastrocnemius and soleus muscles could attenuate the exaggerated ankle muscles' activity (Figure 21) without increasing thigh muscles' activity, enabling the application of different types of EMG-audio feedback (from one or more calf muscles) to improve muscle efficiency during standing. Even in the case of postural sway closer to heel with respect to standing naturally with the EMG-audio feedback condition, it is expected such shift does not compromise body stability in the standing posture.

In relation to standing at ease, we observed participants succeeded in reducing the level of plantar flexors' activity in both biofeedback conditions with the mean CoP position located closer to the ankle joint (i.e., more posteriorly). One could argue whether such difference was enough to bring the mean CoP position closer to the posterior, functional limit of stability, and threaten body stability during standing. Our results revealed the significant posterior shift of mean CoP position when using the feedback from the medial gastrocnemius-soleus was of about 5% of foot length projection with respect to standing at ease. Such difference likely was not enough to move the mean CoP position away from the midfoot. By measuring the CoP excursions while young subjects performed different leaning postures during standing, previous studies observed that: (i) the anterior and posterior boundaries of functional limit of stability correspond roughly to 80% and 15% of the distance between the heels and the toes; and (ii) the mean CoP position is on average located at 40% of foot length during standing naturally, as identified in this study [118,150]. During the EMG biofeedback conditions tested here, therefore, the mean CoP position was likely maintained around midfoot without compromising body stability during standing. Threats to the body stability could exist depending on the size of postural sway while individuals heard the activity of their calf muscles from the EMG-audio feedback. In the present study, however, we observed that the size of postural sway remained within standing at ease levels regardless of EMG biofeedback condition.

Chapter 7

General conclusions

This PhD thesis attempted to bring physiological information about the relevance and the effects of the use of EMG biofeedback for the improvement of muscle efficiency during standing in the elderly. From the studies "*The spatial distribution of ankle muscles activity discriminates aged from young subjects during standing*" (Chapter 4) and "*Is the attenuation effect on the ankle muscles from the EMG biofeedback generalized to – or compensated by – other lower limb muscles during standing*" (Chapter 6), it was concluded that:

1) Different from young individuals, the elderly activates to a greater extent the medial gastrocnemius in terms of the size of muscle's active region and for a prolonged duration the medial gastrocnemius and tibialis anterior muscles during standing balance (Chapter 4). Then, the elderly tends to stand with a greater muscle effort than young subjects. Given that the reduction of excessive muscle activation may be advantageous for postural control, learning to efficiently activate the postural muscles during standing could be beneficial for the control of standing posture in aged individuals, with potential implication for the prevention of falls.

2) The attenuation effect of EMG-audio biofeedback on the ankle muscles during standing is not compensated by other lower limbs muscles when using the feedback from the medial gastrocnemius and tibialis anterior muscles (Chapter 6). These results prove the attenuation effect of EMG biofeedback on the activity of lower limb muscles during standing. Then, the training based on the EMG-audio feedback mainly from the medial gastrocnemius and tibialis anterior muscles might

have a high impact on the rehabilitation of individuals who activate their lower limb muscles excessively during standing, e.g. elderly subjects.

From the methodological studies on differences in the timing of muscle activity, estimated with surface EMGs sampled from a large muscle volume, between calf muscles (Chapters 3 and 5), it was concluded that:

1) The duration of muscle activity differs between and within calf muscles of a single leg. It was observed the medial portion of soleus was almost continuously active while the lateral portion of soleus and the medial portion of gastrocnemius muscle were active more intermittently during standing (Chapter 3). These results suggest high-density surface EMG discriminated well the activity between ankle muscles (i.e., medial gastrocnemius and soleus), as observed from intramuscular EMGs in literature, and muscle activity sampled from different regions of a single muscle (e.g., soleus) can provide estimates more representative of muscle activity during standing. Therefore, key information on the postural control in different populations could be extracted by sampling surface EMGs from a representative muscle volume and from different calf muscles.

2) The timing of calf muscles' activity differs between lower limbs during standing. Specifically, it was demonstrated soleus and medial gastrocnemius were active for different durations between legs (Chapter 5). This finding indicates surface EMGs collected unilaterally may provide a biased view of calf muscles' activity while individuals stand at ease. Thus, surface EMGs should be collected bilaterally to provide a representative estimation of calf muscles' activity in the standing posture. Such result has direct implications on the application of EMG biofeedback for the training of postural control, suggesting a representative feedback of ankle muscles' activity requires the sampling of surface EMGs from both legs.

References

1. Shumway-Cook A, Woollacott MH. *Motor Control: Translating Research into Clinical Practice*. LWW, 2011.

2. Gatev P, Thomas S, Kepple T, et al. Feedforward ankle strategy of balance during quiet stance in adults. *J Physiol* 1999; 514: 915–928.

3. Day B, Steiger M, Thompson P, et al. Effect of vision and stance width on human body motion when standing: implications for afferent control of lateral sway. *J Physiol* 1993; 469: 479–499.

4. Winter DA, Patla AE, Prince F, et al. Stiffness control of balance in quiet standing. *J Neurophysiol* 1998; 80: 1211–1221.

5. Casadio M, Morasso PG, Sanguineti V. Direct measurement of ankle stiffness during quiet standing: Implications for control modelling and clinical application. *Gait Posture* 2005; 21: 410–424.

6. Loram ID, Lakie M. Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for stability. *J Physiol* 2002; 545: 1041–1053.

7. Joseph J, Nightingale A, Williams PL. A detailed study of the electric potentials recorded over some postural muscles while relaxed and standing. *J Physiol* 1955; 127: 617–625.

8. Merletti R, Farina D. *Surface Electromyography: Physiology, Engineering and Applications*. IEEE Press. Wiley, 2016.

9. Merletti R, Farina D. Analysis of intramuscular electromyogram signals. *Philos Trans R Soc A Math Phys Eng Sci* 2009; 367: 357–368.

10. Cavalcanti Garcia MA, Vieira TMM. Surface electromyography: Why, when and how to use it. *Rev Andal Med Deport* 2011; 4: 17–28.

11. Benjuya N, Melzer I, Kaplanski J. Aging-induced shifts from a reliance on sensory input to muscle cocontraction during balanced standing. *J Gerontol A Biol Sci Med Sci* 2004; 59: M166–M171.

12. Baudry S, Lecoeuvre G, Duchateau J. Age-related changes in the behavior of the muscle-tendon unit of the gastrocnemius medialis during upright stance. *J Appl Physiol* 2012; 112: 296–304.

13. Nagai K, Yamada M, Uemura K, et al. Differences in muscle coactivation during postural control between healthy older and young adults. *Arch Gerontol Geriatr* 2011; 53: 338–343.

14. Laughton CA, Slavin M, Katdare K, et al. Aging, muscle activity, and balance control: Physiologic changes associated with balance impairment. *Gait Posture* 2003; 18: 101–108.

15. Kouzaki M, Shinohara M. Steadiness in plantar flexor muscles and its relation to postural sway in young and elderly adults. *Muscle and Nerve* 2010; 42: 78–87.

16. Hermens HJ. Development of recommendations for SEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol* 2000; 10: 361–374.

17. Vieira TMM, Loram ID, Muceli S, et al. Postural activation of the human medial gastrocnemius muscle: are the muscle units spatially localised? *J Physiol* 2011; 589: 431–443.

18. Oliveira VB De, Carneiro SP, Oliveira LF De. Reliability of biceps femoris and semitendinosus muscle architecture measurements obtained with ultrasonography. *Res Biomed Eng* 2016; 32: 365–371.

19. Kellis E, Galanis N, Natsis K, et al. Muscle architecture variations along the human semitendinosus and biceps femoris (long head) length. *J Electromyogr Kinesiol* 2010; 20: 1237–1243.

20. Hodson-Tole EF, Loram ID, Vieira TMM. Myoelectric activity along human gastrocnemius medialis: different spatial distributions of postural and electrically elicited surface potentials. *J Electromyogr Kinesiol* 2013; 23: 43–50.

21. Vieira T, Muceli S, Farina D, et al. Large inter-electrode distances lead to more representative bipolar EMGs, not necessarily affected by crosstalk. *Gait Posture* 2016; 49: S28–S29.

22. Staudenmann D, Kingma I, Daffertshofer A, et al. Heterogeneity of muscle activation in relation to force direction: A multi-channel surface electromyography study on the triceps surae muscle. *J Electromyogr Kinesiol* 2009; 19: 882–895.

23. Reffad A, Mebarkia K, Vieira TMM, et al. Effect of contraction force and knee joint angle on the spatial representation of soleus activity using high-density surface EMG. *Biomed Tech* 2014; 59: 399–411.

24. Holtermann A, Søaard K, Christensen H, et al. The influence of biofeedback training on trapezius activity and rest during occupational computer work: A randomized controlled trial. *Eur J Appl Physiol* 2008; 104: 983–989.

25. Hermens HJ, Hutten MM. Muscle activation in chronic pain: its treatment using a new approach of myofeedback. *Int J Ind Ergon* 2002; 30: 325–336.

26. Heroux ME, Dakin CJ, Luu BL, et al. Absence of lateral gastrocnemius activity and differential motor unit behavior in soleus and medial gastrocnemius during standing balance. *J Appl Physiol* 2014; 116: 140–148.

27. Vieira TMM, Loram ID, Muceli S, et al. Recruitment of motor units in the medial gastrocnemius muscle during human quiet standing: is recruitment intermittent? What triggers recruitment? *J Neurophysiol* 2012; 107: 666–676.

28. Loram ID, Maganaris CN, Lakie M. Human postural sway results from frequent, ballistic bias impulses by soleus and gastrocnemius. *J Physiol* 2005; 564: 295–311.

29. Bottaro A, Yasutake Y, Nomura T, et al. Bounded stability of the quiet standing posture: An intermittent control model. *Hum Mov Sci* 2008; 27: 473–495.

30. Loram ID, Lakie M, Di Giulio I, et al. The consequences of short-range stiffness and fluctuating muscle activity for proprioception of postural joint rotations: the relevance to human standing. *J Neurophysiol* 2009; 102: 460–74.

31. Cabrera JL, Milton JG. On-off intermittency in a human balancing task. *Phys Rev Lett* 2002; 89: 158702.

32. Kiemel T, Zhang Y, Jeka JJ. Identification of neural feedback for upright stance in humans: stabilization rather than sway minimization. *J Neurosci* 2011; 31: 15144–53.

33. Di Giulio I, Maganaris CN, Baltzopoulos V, et al. The proprioceptive and agonist roles of gastrocnemius, soleus and tibialis anterior muscles in maintaining human upright posture. *J Physiol* 2009; 587: 2399–416.

34. Masani K, Sayenko DG, Vette AH. What triggers the continuous muscle activity during upright standing? *Gait Posture* 2013; 37: 72–77.

35. Baudry S, Penzer F, Duchateau J. Input-output characteristics of soleus homonymous Ia afferents and corticospinal pathways during upright standing differ between young and elderly adults. *Acta Physiol* 2014; 210: 667–677.

36. Loram ID, Gollee H, Lakie M, et al. Human control of an inverted pendulum: is continuous control necessary? Is intermittent control effective? Is intermittent control physiological? *J Physiol* 2011; 589: 307–324.

37. Bruce HA, Fredenck PR, Marsha EW. The cost and frequency of hospitalization for fall-related injuries in older adults. *Public Heal briefs* 1992; 82: 5–8.

38. Hartholt KA, Polinder S, Van der Cammen TJM, et al. Costs of falls in an ageing population: A nationwide study from the Netherlands (2007–2009). *Injury* 2012; 43: 1199–1203.

39. Stel VS, Smit JH, Pluijm SMF, et al. Consequences of falling in older men and women and risk factors for health service use and functional decline. *Age Ageing* 2004; 33: 58–65.

40. Talbot LA, Musiol RJ, Witham EK, et al. Falls in young, middle-aged and older community dwelling adults: perceived cause, environmental factors and injury. *BMC Public Health* 2005; 5: 86.

41. Gazibara T, Kurtagic I, Kisic-Tepavcevic D, et al. Falls, risk factors and fear of falling among persons older than 65 years of age. *Psychogeriatrics* 2017; 1–9.

42. Cattagni T, Scaglioni G, Laroche D, et al. Ankle muscle strength discriminates fallers from non-fallers. *Front Aging Neurosci* 2014; 6: 1–7.

43. Lin S-I, Woollacott MH. Postural muscle responses following changing balance threats in young, stable older, and unstable older adults. *J Mot Behav* 2002; 34: 37–44.

44. Woollacott MH, Shumway-Cook A, Nashner LM. Aging and posture control: changes in sensory organization and muscular coordination. *Int J Aging Hum Dev* 1986; 23: 97–114.

45. Woollacott M. Systems contributing to balance disorders in older adutls. *J Gerontol* 2000; 55: 424–428.

46. Horak FB, Shupert CL, Mirka A. Components of postural dyscontrol in the elderly: A review. *Neurobiol Aging* 1989; 10: 727–738.

47. Visser JE, Carpenter MG, van der Kooij H, et al. The clinical utility of posturography. *Clin Neurophysiol* 2008; 119: 2424–2436.

48. Nardone A, Schieppati M. The role of instrumental assessment of balance in clinical decision making. *Eur J Phys Rehabil Med* 2010; 46: 221–237.

49. Taube W, Gruber M, Gollhofer A. Spinal and supraspinal adaptations associated with balance training and their functional relevance. *Acta Physiol* 2008; 193: 101–116.

50. Gruber M, Gruber S, Taube W, et al. Differential effects of ballistic versus sensorimotor training on rate of force development and neural activation in humans. *J Strength Cond Res* 2007; 21: 274–282.

51. Gauffin H, Tropp H, Odenrick P. Effect of ankle disk training on postural control in patients with functional instability of the ankle joint. *Int J Sports Med* 1988; 9: 141–144.

52. Keller M, Pfusterschmied J, Buchecker M, Müller E TW. Improved postural control after slackline training is accompanied by reduced H-reflexes. *Scand J Med Sci Sport* 2012; 22: 471–477.

53. Penzer F, Duchateau J, Baudry S. Effects of short-term training combining strength and balance exercises on maximal strength and upright standing steadiness in elderly adults. *Exp Gerontol* 2015; 61: 38–46.

54. Taube W, Kullmann N, Leukel C, et al. Differential reflex adaptations following sensorimotor and strength training in young elite athletes. *Int J Sports Med* 2007; 28: 999–1005.

55. Gruber M, Gollhofer A. Impact of sensorimotor training on the rate of force

development and neural activation. *Eur J Appl Physiol* 2004; 92: 98–105.

56. Nagai K, Yamada M, Tanaka B, et al. Effects of balance training on muscle coactivation during postural control in older adults: a randomized controlled trial. *Journals Gerontol Ser A Biol Sci Med Sci* 2012; 67: 882–889.

57. Jahnke R, Larkey L, Rogers C, et al. A comprehensive review of health benefits of qigong and tai chi. *Am J Health Promot* 2010; 24: e1–e25.

58. Jeter PE, Nkodo A-F, Moonaz SH, et al. A systematic review of yoga for balance in a healthy population. *J Altern Complement Med* 2014; 20: 221–232.

59. S. Wolf HB, N. Kutner EM, Coogler C, et al. Reducing frailty and falls in older persons: an investigation of Tai Chi and computerized balance training. Atlanta FICSIT Group. Frailty and Injuries: Cooperative Studies of Intervention Techniques. *J Am Geriatr Soc* 1996; 44: 489–497.

60. Li F, Harmer P, Fisher KJ, et al. Tai Chi and fall reductions in older adults: a randomized controlled trial. *J Gerontol A Biol Sci Med Sci* 2005; 60: 187–194.

61. Loftus SL. Qi Gong to improve postural stability (QTIPS) for parkinson fall prevention. *Top Geriatr Rehabil* 2014; 30: 58–69.

62. Society AG, Society BG. Guideline for the prevention of falls in older persons. American Geriatrics Society, British Geriatrics Society, and American Academy of Orthopaedic Surgeons Panel on Falls Prevention. *J Am Geriatr Soc* 2001; 49: 664–72.

63. Chiari L, Dozza M, Cappello A, et al. Audio-biofeedback for balance improvement: An accelerometry-based system. *IEEE Trans Biomed Eng* 2005; 52: 2108–2111.

64. Dault MC, de Haart M, Geurts ACH, et al. Effects of visual center of pressure feedback on postural control in young and elderly healthy adults and in stroke patients. *Hum Mov Sci* 2003; 22: 221–236.

65. Cawsey RP, Chua R, Carpenter MG, et al. To what extent can increasing the magnification of visual feedback of the centre of pressure position change the control of quiet standing balance? *Gait Posture* 2009; 29: 280–284.

66. Van Peppen R, Kortsmit M, Lindeman E, et al. Effects of visual feedback therapy on postural control in bilateral standing after stroke: a systematic review. *J Rehabil Med* 2006; 38: 3–9.

67. Dos Anjos F, Lemos T, Imbiriba LA. Does the type of visual feedback information change the control of standing balance? *Eur J Appl Physiol* 2016; 116: 1771–1779.

68. Zijlstra GA, Mancini M, Chiari L, et al. Biofeedback for training balance and mobility tasks in older populations: a systematic review. *J Neuroeng Rehabil* 2010; 7: 58.

69. Mirelman A, Herman T, Nicolai S, et al. Audio-biofeedback training for posture and balance in patients with Parkinson's disease. *J Neuroeng Rehabil* 2011; 8: 35.

70. Dozza M, Chiari L, Horak FB. Audio-biofeedback improves balance in patients with bilateral vestibular loss. *Arch Phys Med Rehabil* 2005; 86: 1401–1403.

71. Sihvonen S, Sipilä S, Taskinen S, et al. Fall incidence in frail older women after individualized visual feedback-based balance training. *Gerontology* 2004; 50: 411–416.

72. Vieira TM, Baudry S, Botter A. Young, healthy subjects can reduce the activity of calf muscles when provided with EMG biofeedback in upright stance. *Front Physiol* 2016; 7: 1–12.

73. Giggins OM, Persson UM, Caulfield B. Biofeedback in rehabilitation. *J Neuroeng Rehabil* 2013; 10: 60.

74. Wright ZA, Rymer WZ, Slutzky MW. Reducing abnormal muscle coactivation after stroke using a myoelectric-computer interface: a pilot study. *Neurorehabil Neural Repair* 2014; 28: 443–51.

75. Nouwen A. EMG biofeedback used to reduce standing levels of paraspinal muscle tension in chronic low back pain. *Pain* 1983; 17: 353–360.

76. Nestoriuc Y, Martin A, Rief W, et al. Biofeedback treatment for headache disorders: a comprehensive efficacy review. *Appl Psychophysiol Biofeedback* 2008; 33: 125–140.

77. Basmajian J V. Control and training of individual motor units. *Science* 1963; 141: 440–1.

78. Huang H, Wolf SL, He J. Recent developments in biofeedback for neuromotor rehabilitation. *J Neuroeng Rehabil* 2006; 3: 11.

79. Ma EP-M, Yiu GK-Y, Yiu EM-L. The effects of self-controlled feedback on

learning of a ‘relaxed phonation task’. *J Voice* 2013; 27: 723–8.

80. Ng GYF, Zhang AQ, Li CK. Biofeedback exercise improved the EMG activity ratio of the medial and lateral vasti muscles in subjects with patellofemoral pain syndrome. *J Electromyogr Kinesiol* 2008; 18: 128–133.

81. Doan-Aslan M, Nakipolu-Yüzer GF, Doan A, et al. The effect of electromyographic biofeedback treatment in improving upper extremity functioning of patients with hemiplegic stroke. *J Stroke Cerebrovasc Dis* 2012; 21: 187–192.

82. Basmajian J V. Research foundations of EMG biofeedback in rehabilitation. *Biofeedback Self Regul* 1988; 13: 275–298.

83. Wolf SL. Electromyographic biofeedback applications to stroke patients. A critical review. *Phys Ther* 1983; 63: 1448–1459.

84. Stanton R, Ada L, Dean CM, et al. Biofeedback improves activities of the lower limb after stroke: a systematic review. *J Physiother* 2011; 57: 145–155.

85. Glombiewski JA, Bernardy K, Häuser W. Efficacy of EMG- and EEG-biofeedback in fibromyalgia syndrome: a meta-analysis and a systematic review of randomized controlled trials. *Evidence-based Complement Altern Med*; 2013.

86. Dannecker C, Wolf V, Raab R, et al. EMG-biofeedback assisted pelvic floor muscle training is an effective therapy of stress urinary or mixed incontinence: a 7-year experience with 390 patients. *Arch Gynecol Obstet* 2005; 273: 93–97.

87. Wilkinson M. Biofeedback in General Relaxation Training. *Arch Phys Med* 1975; 11: 182–184.

88. Poppen R, Hanson HB, Ip SM. Generalization of EMG biofeedback training. *Biofeedback Self Regul* 1988; 13: 235–43.

89. Ince LP, Leon MS, Christidis D. Experimental foundations of EMG biofeedback with the upper extremity: a review of the literature. *Biofeedback Self Regul* 1984; 9: 371–83.

90. Ma C, Szeto GP, Yan T, et al. Comparing biofeedback with active exercise and passive treatment for the management of work-related neck and shoulder pain: A randomized controlled trial. *Arch Phys Med Rehabil* 2011; 92: 849–858.

91. LeVeau BF, Rogers C. Selective training of the vastus medialis muscle using EMG biofeedback. *Phys Ther* 1980; 60: 1410–5.

92. Christanell F, Hoser C, Huber R, et al. The influence of electromyographic biofeedback therapy on knee extension following anterior cruciate ligament reconstruction: a randomized controlled trial. *Sports Med Arthrosc Rehabil Ther Technol* 2012; 4: 41.

93. Vedsted P, Søgaard K, Blangsted AK, et al. Biofeedback effectiveness to reduce upper limb muscle activity during computer work is muscle specific and time pressure dependent. *J Electromyogr Kinesiol* 2011; 21: 49–58.

94. Palmerud G, Sporrong H, Herberts P, et al. Consequences of trapezius relaxation on the distribution of shoulder muscle forces: An electromyographic study. *J Electromyogr Kinesiol* 1998; 8: 185–193.

95. Horak FB, Frank J, Nutt J. Effects of dopamine on postural control in parkinsonian subjects: scaling, set, and tone. *J Neurophysiol* 1996; 75: 2380–2396.

96. Basmajian J V., De Luca CJ. *Muscles Alive: their functions revealed by electromyography*. 5th Editio. Williams & Wilkins, 1985.

97. Winter DA. Human balance and posture control during standing and walking. *Gait and Posture* 1995; 3: 193–214.

98. Loram ID, Maganaris CN, Lakie M. Paradoxical muscle movement in human standing. *J Physiol* 2004; 556: 683–689.

99. Joseph J, Nightingale A. Electromyography of muscles of posture: leg and thigh muscles in women, including the effects of high heels. *J Physiol* 1956; 132: 465–8.

100. Mochizuki G, Semmler JG, Ivanova TD, et al. Low-frequency common modulation of soleus motor unit discharge is enhanced during postural control in humans. *Exp Brain Res* 2006; 175: 584–595.

101. Vieira TMM, Windhorst U, Merletti R. Is the stabilization of quiet upright stance in humans driven by synchronized modulations of the activity of medial and lateral gastrocnemius muscles? *J Appl Physiol* 2010; 108: 85–97.

102. Loram ID, Lakie M. Human balancing of an inverted pendulum: position control by small, ballistic-like, throw and catch movements. *J Physiol* 2002; 540: 1111–1124.

103. Merletti R, Botter A, Troiano A, et al. Technology and instrumentation for detection and conditioning of the surface electromyographic signal: state of the art. *Clinical Biomechanics* 2009; 24: 122–134.

104. Loram ID, Maganaris CN, Lakie M. Paradoxical muscle movement in human standing. *J Physiol* 2004; 556: 683–689.

105. Edgerton VR, Smith JL, Simpson DR. Muscle fibre type populations of human leg muscles. *Histochem J* 1975; 7: 259–266.

106. Merletti R, Parker P (eds). *Electromyography: Physiology, Engineering, and Noninvasive Applications*. Hoboken, NJ, USA: John Wiley & Sons, Inc. Epub ahead of print 12 July 2004. DOI: 10.1002/0471678384.

107. Jacono M, Casadio M, Morasso PG, et al. The sway-density curve and the underlying postural stabilization process. *Motor Control* 2004; 8: 292–311.

108. Malisoux L, Francaux M, Theisen D. What do single-fiber studies tell us about exercise training? *Med Sci Sports Exerc* 2007; 39: 1051–1060.

109. Simoneau JA, Bouchard C. Human variation in skeletal muscle fiber-type proportion and enzyme activities. *Am J Physiol* 1989; 257: E567-72.

110. Henry SM, Fung J, Horak FB. EMG responses to maintain stance during multidirectional surface translations. *J Neurophysiol* 1998; 80: 1939–1950.

111. Vieira TMM, Minetto MA, Hodson-Tole EF, et al. How much does the human medial gastrocnemius muscle contribute to ankle torques outside the sagittal plane? *Hum Mov Sci* 2013; 32: 753–767.

112. Loram ID, Lakie M, Gawthrop PJ. Visual control of stable and unstable loads: what is the feedback delay and extent of linear time-invariant control? *J Physiol* 2009; 587: 1343–65.

113. Brown JMM, Wickham JB, McAndrew DJ, et al. Muscles within muscles: coordination of 19 muscle segments within three shoulder muscles during isometric motor tasks. *J Electromyogr Kinesiol* 2007; 17: 57–73.

114. Farina D, Leclerc F, Arendt-Nielsen L, et al. The change in spatial distribution of upper trapezius muscle activity is correlated to contraction duration. *J Electromyogr Kinesiol* 2008; 18: 16–25.

115. McLean L, Goudy N. Neuromuscular response to sustained low-level muscle activation: within- and between-synergist substitution in the triceps surae muscles. *Eur J Appl Physiol* 2004; 91: 204–216.

116. Tomioka K, Iwamoto J, Saeki K, et al. Reliability and validity of the international physical activity questionnaire (IPAQ) in elderly adults: the fujiwara-kyo study. *J Epidemiol* 2011; 21: 459–465.

117. Chiari L, Rocchi L, Cappello A. Stabilometric parameters are affected by anthropometry and foot placement. *Clin Biomech* 2002; 17: 666–677.

118. Duarte M, Zatsiorsky VM. Effects of body lean and visual information on the equilibrium maintenance during stance. *Exp Brain Res* 2002; 146: 60–69.

119. Pinsault N, Vuillerme N. Test-retest reliability of centre of foot pressure measures to assess postural control during unperturbed stance. *Med Eng Phys* 2009; 31: 276–286.

120. Gallina A, Pollock CL, Vieira TM, et al. Between-day reliability of triceps surae responses to standing perturbations in people post-stroke and healthy controls: A high-density surface EMG investigation. *Gait Posture* 2016; 44: 103–109.

121. Vieira TMM, Merletti R, Mesin L. Automatic segmentation of surface EMG images: improving the estimation of neuromuscular activity. *J Biomech* 2010; 43: 2149–2158.

122. Farina D, Cescon C, Merletti R. Influence of anatomical, physical, and detection-system parameters on surface EMG. *Biol Cybern* 2002; 86: 445–456.

123. Onambele GL, Narici M V, Maganaris CN. Calf muscle-tendon properties and postural balance in old age. *J Appl Physiol* 2006; 100: 2048–2056.

124. Avancini C, De Oliveira LF, Menegaldo LL, et al. Variations in the spatial distribution of the amplitude of surface electromyograms are unlikely explained by changes in the length of medial gastrocnemius fibres with knee joint angle. *PLoS One* 2015; 10: e0126888.

125. Oliveira LF, Simpson DM, Nadal J. Calculation of area of stabilometric signals using principal component analysis. *Physiol Meas* 1996; 17: 305–12.

126. Warnica MJ, Weaver TB, Prentice SD, et al. The influence of ankle muscle activation on postural sway during quiet stance. *Gait Posture* 2014; 39: 1115–1121.

127. Mesin L, Merletti R, Vieira TMM. Insights gained into the interpretation of surface electromyograms from the gastrocnemius muscles: a simulation study. *J Biomech* 2011; 44: 1096–1103.

128. Blouin JS, Brown HJ, Héroux ME, et al. Reply from J. S. Blouin, H. J. Brown, M. E. Héroux, J. T. Inglis and G. P. Siegmund. *J Physiol* 2016; 594: 1955–1955.

129. Vieira TM, Wakeling JM, Hodson-Tole EF. Is there sufficient evidence to claim muscle units are not localised and functionally grouped within the human gastrocnemius? *J Physiol* 2016; 594: 1953–1954.

130. Amiridis IG, Hatzitaki V, Arabatzi F. Age-induced modifications of static postural control in humans. *Neurosci Lett* 2003; 350: 137–140.

131. Sayenko D, Alekhina M, Masani K, et al. Positive effect of balance training with visual feedback on standing balance abilities in people with incomplete spinal cord injury. *Spinal Cord* 2010; 48: 886–893.

132. Horak FB, Henry SM, Shumway-Cook a. Postural perturbations: new insights for treatment of balance disorders. *Phys Ther* 1997; 77: 517–533.

133. Horak FB, Nashner LM. Central programming of postural movements: adaptation to altered support-surface configurations. *J Neurophysiol* 1986; 55: 1369–1381.

134. Sozzi S, Nardone A, Schieppati M. Calibration of the leg muscle responses elicited by predictable perturbations of stance and the effect of vision. *Front Hum Neurosci* 2016; 10: 419.

135. Peterka RJ. Postural control model interpretation of stabilogram diffusion analysis. *Biol Cybern* 2000; 82: 335–343.

136. Masani K, Popovic MR, Nakazawa K, et al. Importance of body sway velocity information in controlling ankle extensor activities during quiet stance. *J Neurophysiol* 2003; 90: 3774–82.

137. Marsden JF, Castellote J, Day BL. Bipedal distribution of human vestibular-evoked postural responses during asymmetrical standing. *J Physiol* 2002; 542: 323–331.

138. Wu G, MacLeod M. The control of body orientation and center of mass location under asymmetrical loading. *Gait Posture* 2001; 13: 95–101.

139. Mochizuki G, Ivanova TD, Garland SJ. Factors affecting the common modulation of bilateral motor unit discharge in human soleus muscles. *J Neurophysiol* 2007; 97: 3917–3925.

140. Monsour M, Ivanova TD, Wilson TD, et al. Influence of vestibular afferent input on common modulation of human soleus motor units during standing. *Motor Control* 2012; 16: 466–79.

141. Duarte M, Zatsiorsky VM. Patterns of center of pressure migration during prolonged unconstrained standing. *Motor Control* 1999; 3: 12–27.

142. Liang S, Xu J, wang L, et al. An investigation into the bilateral functional differences of the lower limb muscles in standing and walking. *PeerJ* 2016; 4: e2315.

143. Dos Anjos F V., Pinto TP, Gazzoni M, et al. The spatial distribution of ankle muscles activity discriminates aged from young subjects during standing. *Front Hum Neurosci* 2017; 11: 1–12.

144. Dawson-Saunders B, Trapp RG. *Basic & Clinical Biostatistics*. Second edi. Appleton & Lange, 1994.

145. Haddad JM, Rietdyk S, Ryu JH, et al. Postural asymmetries in response to holding evenly and unevenly distributed loads during self-selected stance. *J Mot Behav* 2011; 43: 345–355.

146. Blaszczyk JW, Prince F, Raiche M, et al. Effect of ageing and vision on limb load asymmetry during quiet stance. *J Biomech* 2000; 33: 1243–1248.

147. Lee SSM, Piazza SJ. Inversion-eversion moment arms of gastrocnemius and tibialis anterior measured in vivo. *J Biomech* 2008; 41: 3366–3370.

148. Genthon N, Gissot AS, Froger J, et al. Posturography in patients with stroke: estimating the percentage of body weight on each foot from a single force platform. *Stroke* 2008; 39: 489–491.

149. Agur AM, Ng-Thow-Hing V, Ball KA, et al. Documentation and three-dimensional modelling of human soleus muscle architecture. *Clin Anat* 2003; 16: 285–293.

150. Schieppati M, Hugon M, Grasso M, et al. The limits of equilibrium in young and elderly normal subjects and in parkinsonians. *Electroencephalogr Clin Neurophysiol* 1994; 93: 286–298.