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Section I – Kinesiology

Could Ankle Muscle Activation Be Used as a Simple Measure of Balance Exercise Intensity?

by
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Few, if any, studies have reported the effects of intensity of balance exercise for balance training and rehabilitation. The aim of the present study was to find a relative measure of intensity of balance exercise. On this basis, we analysed ankle muscle activation in the sagittal plane with increasing difficulty for a one leg stance on a T-board. Ten adults (7 men, 24.1 ± 3.5 years; 3 women, 30.6 ± 5.8 years) performed 3 trials on a T-board within 6 randomly assigned stability levels. T-board swaying velocities in the sagittal plane were manipulated to attain different stability levels (conditions). Concurrently, angular distance of the T-board and active balance time (i.e., percentage of a total time balancing) under each condition were measured. Surface electromyography from the tibialis anterior, gastrocnemius and soleus were monitored during one leg stance. The surface electromyography amplitude in the time domain was quantified using the root-mean-square values. Significant effect of stability levels on angular distance ($F_{5,45}$ = 3.4; p = 0.01) and velocity of the T-board ($F_{5,45}$ = 4.6; p = 0.002) were obtained. Active balance time decreased by ~15% (p = 0.001) from the maximal to the minimal stability conditions. The graded level of balance board stability conditions did not generate significantly higher root-mean-square values in any muscles and hence could not be used as a relative measure of intensity of balance exercise. These findings imply that there could be a plateau in difficulty of balance exercise for enhancement of ankle muscle activity.

Key words: *soleus, tibialis anterior, lateral gastrocnemius, EMG.*

Introduction

Balance exercise generally encompasses all forms of exercise that challenge the postural control system. Consequently, such exercises have been implemented with the primary intention to increase postural stability, a requirement for maintaining steadiness during static and dynamic activities (Wikstrom, 2005). For improved balance performance outcomes, balance exercises must be dosed appropriately across the variables of frequency, intensity, type and duration (Phillips and Kennedy, 2012; Thompson et al., 2010). It has been already shown that aerobic and strength training requires a minimum threshold in order to be effective. The American College Sport Medicine – ACSM (2014) notes that muscle-strengthening activities should be done at high levels of intensity and for 2-3 sessions per week, while, for aerobic exercise, minimum effective dose is 150 to 300 minutes per week of moderate-intensity or 75 to 150 min per week of physical activity at vigorous intensity. On the other hand, balance exercise recommendations only rely on a frequency prescription (i.e., 2-3 sessions per week), type (static, dynamic exercise) and time (45-60 min) spent exercising, without prescribing exercise intensity (Espy et al., 2017; Farlie et al., 2018; Lesinski et al., 2015). Intensity describes how hard the body needs to work to perform a given activity (Phillips and Kennedy, 2012). Thus, it is also an important factor to consider when prescribing balance exercising. The principle of overloading dictates that exercise needs to be performed at or near the limits of individual maximal capacity to induce a training effect

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(Thompson et al., 2010). Since, individuals have unique balance capabilities, quantifying an outcome measure of intensity of balance exercise still remains crucial. A balance exercise prescription based only on variables of frequency, type and duration may not be sufficient to improve postural control. Since a unique measure of balance exercise intensity does not exist (Farlie et al., 2013, 2018; Haas et al., 2012; Lubetzky-Vilnai et al., 2010; Maszczyk et al., 2018), there remains a need for an efficient measure capable of rating the intensity of balance exercise. Indeed, a very recent systematic review by Farlie et al. (2018) also revealed that frequency, time and type of balance exercises reported in interventions to improve postural control in the elderly accounted only for ~2% of the variance in balance performance outcomes. Although, this was the first known attempt at using meta-regression to analyse the relative contribution of reported exercise prescription variables (i.e., frequency, time and type of activity) on balance performance, it seems that previously mentioned exercise prescription variables are insufficient for optimal progression and prescription of balance exercises (Farlie et al., 2018).

In general, intensity of exercise can be expressed in relative (e.g., percentage of maximal oxygen uptake or 1-repetition maximum for a given activity) or absolute measures (e.g., work for, or total load for, a given activity). Relative measures of intensity might be a better indicator to determine the appropriate intensity for a given exercise than absolute measures (Phillips and Kennedy, 2012). With this in mind, measurement of intensity of balance exercise in relative terms may be viewed as a percentage of an individual's maximal capacity of the postural control system (i.e., a common name for the complex interaction of the musculoskeletal and neural system to control our body in space). A direct measurement of maximal capacity of the postural control system has not been established to date, since postural control is not regulated by a single system, i.e., a motor system, but emerges from the interaction of at least two additional systems, e.g. sensory and cognitive (Shumway-Cook and Woollacott, 2017). Therefore, the most common approach being measured is body sway excursion during balancing in different positions (Billot et al., 2010; Lemos et al., 2015; Warnica et al., 2014). Occurrences of body sway during balance exercise are usually evaluated through the variables of the centre of the distribution of the total force applied to the supporting surface - CoP (Shumway-Cook and Woollacott, 2017). It has been shown that total CoP excursions significantly increased with increased difficulty of balance exercise, when the base of support (i.e., from bipedal to unipedal stance, or from wide to shorter stance width) and the use of sensory information were gradually reduced (i.e., from firm to foam ground or from open to closed eyes) (Lemos et al., 2015; Muehlbauer et al., 2012). It can be argued, therefore, that this approach i.e., classifying difficulty of balance exercise with respect to CoP variables, is sufficient for optimal balance exercise prescription and progression. However, as with other absolute measures of exercise intensity, this approach is based on the assumption that all individuals have the same balance capacity and are consistently challenged by the introduction of a subsequent exercise in the hierarchy (Farlie et al., 2013). In fact, on the basis of total CoP displacement, we might conclude that a two-leg stance is easier than one leg stance (Muehlbauer et al., 2012), but from this point of view, it is not clear how much smaller challenge to an individual's balance capabilities would be. Secondly, a decrease in CoP displacement does not always reflect improvements in postural control. For example, during instances where balance is threatened, older people constrict sway area to prevent the CoP from deviating towards postural extremes about the base of support (Horak et al., 1992; Lee et al., 2015). On the other hand, some older adults use larger and higher-frequency excursions of the CoP to enhance sensory information about their posture while remaining well within their limits of stability (Patla et al., 1990).

One leg stances on stable and unstable surfaces are often integrated in balance interventions (Zech et al., 2009, 2010). For body sway corrections while standing on a one-leg stance, healthy adults predominately activate muscles encompassing the ankle joint (Muehlbauer et al., 2014). Moreover, with increasing one leg stance difficulty (i.e., from eyes open, firm ground over eyes open, foam ground to eyes closed firm ground) postural sway significantly increased (Muehlbauer et al., 2014) as did the amplitude of the surface electromyographic signal (EMG) of ankle muscles (e.g., tibialis anterior EMG amplitude increased from ~60 ŋV with eyes open to ~100 ŋV with eyes closed on a stable ground, and from ~80 ŋV with eyes open to ~120 ŋV with eyes closed on a balance platform) (Braun Ferreira et al., 2011). Since the maximal EMG for a given muscle is easy to measure and well established, the percentage of maximal EMG amplitude may also be used as an outcome for the relative measure of balance exercise intensity (Lesinski et al., 2015). Specifically, Braun Ferreira et al. (2011) revealed that ankle muscle activity (i.e., tibialis anterior, tibialis posterior, peroneus longus, medial and lateral gastrocnemius) during a one-leg stance on the majority of unstable surfaces was greater compared to stable surfaces. Furthermore, a significantly greater ankle muscle activity was observed for eyes closed compared to eyes open conditions (except for the gastrocnemius medialis) (Braun Ferreira et al., 2011). Likewise, Muehlbauer et al. (2014) analyzed CoP displacements and activity of four lower leg muscles (i.e., m. tibialis anterior, m. soleus, m. gastrocnemius medialis, m. peroneus longus) during a one-leg stance under various sensory conditions with an increased level of exercise difficulty. An increase in sensory exercise difficulty resulted in deteriorated balance performance and increased muscle activity for all but two muscles (i.e., m. gastrocnemius medialis and m. peroneus longus). Surface instability probably induced unforeseen perturbations, generating rapid changes in the length of the ankle ligament, thereby, greater afferent stimuli and reflexive motor responses in order to produce rapid joint stability (Myers et al., 2003), which may have induced greater muscle activity. Therefore, the obtained increase in EMG could not be related only to increases in postural demands (i.e., greater intensity), but also to the removal of visual control or the diffusion of proprioceptive information, i.e., provoking less predictable information (Hytönen et al., 1993) with focus on reactive postural control. Consequently, it is important to provide a more systematic increase in difficulty levels of balance exercise, without changes in sensory conditions, to establish whether ankle muscle activity systematically increases with increment in balance exercise difficulty. Since the maximal EMG amplitude for a given muscle is easy to measure, the relative measure of balance exercise intensity may be established as well.

To our knowledge, there have been no systematic studies varying the difficulty levels of chosen balance exercises and simultaneous monitoring of ankle muscle activation. Therefore, the objective of this study was to examine the ankle muscle activation (i.e., m. soleus, m. lateral gastrocnemius, and m. tibialis anterior) during a one-leg stance with greater instability and increased difficulty (i.e., increased angular velocity and distance) on a T-board in the sagittal plane and to establish whether the percentage of maximal ankle muscle activation could be used as a relative measure of balance exercise intensity. We hypothesised that an increase in difficulty in the one leg stance would be coupled with an increase in the percentage of maximal ankle muscle activation and could be used as an outcome of relative measure of balance exercise intensity.

Methods

Participants

Ten randomly assigned healthy young adults (7 men, 24.1 ± 3.5 years; 178.7 ± 6.9 cm, 80.2 ± 8.1 kg and 3 women, 30.6 ± 5.8 years, 169 ± 5.1 cm; 61 ± 2.6 kg) participated in the study. They had not taken part in any systematic training for at least one year prior to, and during the experiment. None had any history of musculoskeletal, neurological or orthopaedic disorders that might have affected the results of the experiment. All of them were right-foot dominant as determined by the lateral preference inventory (Coren, 1993). All participants signed an informed written consent form. The study protocol followed the principles of the Declaration of Helsinki and was approved by the University Ethics Committee (Faculty of Sport, University of Ljubljana).

Design and Procedures

Each participant took part in two laboratory sessions: familiarisation and data collection. The familiarisation session was performed 48 to 72 hours before data collection. During the familiarisation session, participants were asked to carry out a one leg stance on a balance board (T-board) to become acquainted with the experimental settings and to minimise the learning effect during data collection. Data collection began with body height and mass measurements, which were followed by preparation of participant's skin and EMG electrode placement on ankle muscles (m. soleus, m. tibialis anterior and m. gastrocnemius). Participants undertook a standardised warm up protocol followed shortly after by ankle muscle maximal voluntary contraction (MVC) measurements (Figure 1). Participants warmed up for 6 min performing alternate stepping on a 20 cm high bench in the rhythm of 0.5 Hz. MVC measurements (dorsal and plantar flexion) were carried out exactly two minutes after the warm up. A one-leg stance on the T-board was performed 5 minutes after the MVC measurements (Figure 1). Participants began with randomly assigned conditions (Con1 to Con6) to avoid habituation and fatigue. The rest period between conditions was exactly two minutes. Test circumstances (e.g., room illumination and noise) were in accordance with the recommendation for postural assessment (Kapteyn et al., 1983).

Measures

The one leg stance was performed on a custom-made balance T-board (Faculty of Sport, Ljubljana) constructed of wood (width: 25 cm, length: 40 cm and height: 7 cm), comprising a flat, wooden surface, placed on a narrow dowel fastened with a hinge to a wooden platform (3 cm) that prevented the balance board from shifting in a different direction and only permitting uniaxial movement of the ankle (i.e., sagittal plane). Foam manipulation was used to achieve the different angular velocity of T-board swaying. The foam was systematically placed (but randomly for each participant) sideways from the left under the board between the board and its platform in its mid-section. The greater the foam length beneath the balance board, the less difficult the stance was, since the angular velocity was decreased. Condition 1 (Con1) represents the slowest (the foam was completely under the board – 25 cm), and the condition 6 (Con6) the fastest angular velocity (the foam was completely removed). For the mid-level conditions (Con2 – Con5), foam was withdrawn by 5 cm to increase the angular velocity of swaying from the second to the fifth setting.

During the balancing task, the stance foot was positioned in the centre of the board, perpendicularly aligned to the dowel on the underside. Participants tried to maintain their equilibrium on the board by means of ankle plantar and dorsal flexions and with the knee held at near to full extension such that neither the posterior nor the anterior edge of the board touched the ground. Balancing on the board was performed with eyes open, and with the knee of the opposite leg held in a semi-flexed position. Prior to starting a balancing task, participants were required to hold a switch button attached to a bar near the board. When the postural equilibrium (i.e., anterior and posterior edge lifted from the floor) had been achieved, the button was released and hands were placed across the chest. The off-switch was the starting signal for recording active balancing time, balance board angular distance and surface EMG activity.

Specifically, participants were asked to activate the switch button as balance was lost (i.e., just before anterior and posterior edge of the T-board touched the ground) and release it, when equilibrium was re-established. Participants were instructed to avoid any delay between the edge of the board making contact with the ground and activating the button switch. They were also instructed to re-establish the equilibrium position as quickly as possible for each trial. The percentage of active time (i.e., equilibrium time - the time spent actively balancing on the T-board was calculated as the difference between the total trial duration (i.e., 20 s) and the duration of the postural equilibrium (both edges of the boards lifted from the floor)) was subjected to further analysis. The angular distance of T-board swaying was measured with a uniaxial goniometer (Biovision, Wehrhaim, Germany), which was embedded into the sagittal plane of the T-board. The following variables were later calculated: (1) average angular distance (°) during active balancing, and (2) average angular velocity ($°s^{-1}$) during active balancing - the ratio between the average angular distance during the trial and active time. The mean of the three trials was used for further statistical analysis. All measurements were taken with a PowerLab A/D system (16/30 - ML880/P) and analysed with LabChart software (both from ADInstruments, Bella Vista, Australia). The EMG signal was amplified with an octal bio-amplifier (ML138, ADInstruments) with a bandwidth frequency ranging from 3 to 1,000 Hz (input impedance = 200 MX, common mode rejection ratio = 85 dB, gain = 1,000).

Maximal isometric voluntary plantar flexion and extension were performed with an isometric ankle torque measuring device (own design), equipped with an isometric dynamometer (MES, Maribor, Slovenia). Measurement required the participants to be in a seated position (ankle, knee and hip angles were at 90 degrees) with hands held onto the device to further stabilize the trunk. The rotational axis of the dynamometer was visually aligned to the rotational axis of the ankle (i.e., lateral malleolus). Neuromuscular test sessions began by three maximal plantar flexions, all brief (~5 s) and separated by ≥ 60 s of rest. At the assistant's verbal signal, the participant gradually increased (over 2 s period) maximal torque to a plateau and then proceeded with smooth maximal isometric force/torque for 3 s. Neuromuscular tests for the ankle dorsal flexion followed exactly the same protocol as for the ankle plantar flexion, and began ≥120 s later. The assistant was loud, clear and gave encouragement to the participant to ensure maximal motivation. The highest value/the average value of three efforts in a single test session was taken for further consideration. The maximal voluntary torque (expressed as Nm) was defined as the maximum values recorded for 0.5 s, when torque had reached a plateau.

EMG measurements. All EMG signals were measured with a pair of bipolar (2.5 cm inter-electrode distance) self-adhesive electrodes (Ag-AgCl, Kendall Tyco Arbo, H124SG, 24 mm). For all ankle muscles (m. soleus, m. tibialis anterior, m. lateral gastrocnemius), the electrodes were positioned according to *SENIAM* recommendations with the reference electrode placed over the lateral malleolus. The participants' skin areas were properly prepared prior to testing. EMG data were recorded with a PowerLab system (16/30 - ML880/P, ADInstruments, Bella Vista, Australia) at a sampling frequency of 1 kHz. The EMG signals were amplified with an octal bio-amplifier (ML138, ADInstruments) with a bandwidth frequency ranging from 3 to 1000 Hz (input impedance = 200 MΩ, common mode rejection ratio = 85 dB, gain = 1000), and analysed with LabChart7 software (ADInstruments). Raw EMG data were filtered (bandpass 10 - 500 Hz). All signals were acquired in microvolts.

The surface EMG amplitude in the time domain was quantified using the root-mean-square (RMS) and processed as a moving average over 300 ms (Tomazin et al., 2016). The RMS EMG values were selected for each one-leg stance on the T-board and maximal isometric ankle dorsal and plantar flexion. Surface EMG signals of the one leg stance were analysed using only the portion of the EMG signal taken during active balancing on the T-board (both board edges were raised from the floor). For maximal isometric voluntary contractions, an interval comprising 0.5 s of the torque plateau was analysed. Therefore, the highest root mean square values of soleus ($_{SO}RMS_{MVC}$), tibialis anterior ($_{TA}RMS_{MVC}$) and gastrocnemius lateralis ($_{GL}RMS_{MVC}$) activation were retained for normalisation of the surface EMG recorded during active balancing on the T-board. Furthermore, RMS EMG data obtained from each one-leg stance were normalized by the RMS EMG of the maximal voluntary contractions during plantar and dorsal ankle flexion [$_{SO}RMS$ (%), $_{TA}RMS$ (%) and $_{GL}RMS$ (%)]. Average values across three trials under each condition were retained for further statistical analyses.

Statistical Analysis

Descriptive statistics was calculated for all data and are represented as mean ± standard deviations (SD). Normal distribution of the presented variables was analysed using the Shapiro – Wilk test. A one-way analysis of variance with repeated measures was performed to reveal the differences between angular distance and velocity of balance board swaying. Statistical comparison for active balancing time and muscle activation between six conditions of balance board swaying was also performed using a one-way repeated measures analysis of variance. In case of significant F values, pairwise differences between mean values were identified using a Fisher LSD post-hoc test. Effect size was calculated using partial eta squared ($\eta 2$), where the values of the effect sizes of 0.01, 0.06 and above 0.14 were considered small, medium, and large, respectively (Field, 2013). The statistical analyses were performed with Statistica 6.0 statistical package (StatSoft, Inc., Tulsa, USA) and SPSS software (version 13.0). The alpha level was set at .05.

Results

All data were normally distributed. Average angular distance (degrees) during active balancing increased from the first two (Con 1 and 2) to the last three conditions (Con 4, 5 and 6; Figure 2A; $F_{5,45}$ = 3.4; *p* = 0.01; η^2 = 0.27). Accordingly, angular velocity of the T-board was greater during the last three compared to first three conditions (Figure 2B; $F_{5,45}$ = 4.6; *p* = 0.002; η^2 = 0.33). Therefore, exercise difficulty was greater for the last three conditions compared to the first three ones since active balancing time on the T-board was shorter (Figure 2C, $F_{5,45}$ = 7.6; *p* = 0.000; η^2 = 0.45).

On the other hand, comparison of ankle muscle activity did not reveal significant effect of condition on SORMS ($F_{5,45}$ = 0.5; *p* = 0.71, η^2 = 0.06; Table 1 and Figure 3A), TARMS ($F_{5,45}$ = 1; *p* = 0.43, η^2 = 0.10; Table 1 and Figure 3B) and GLRMS ($F_{5,45}$ = 1.1; *p* = 0.36, η^2 = 0.11; Table 1 and Figure 3C).

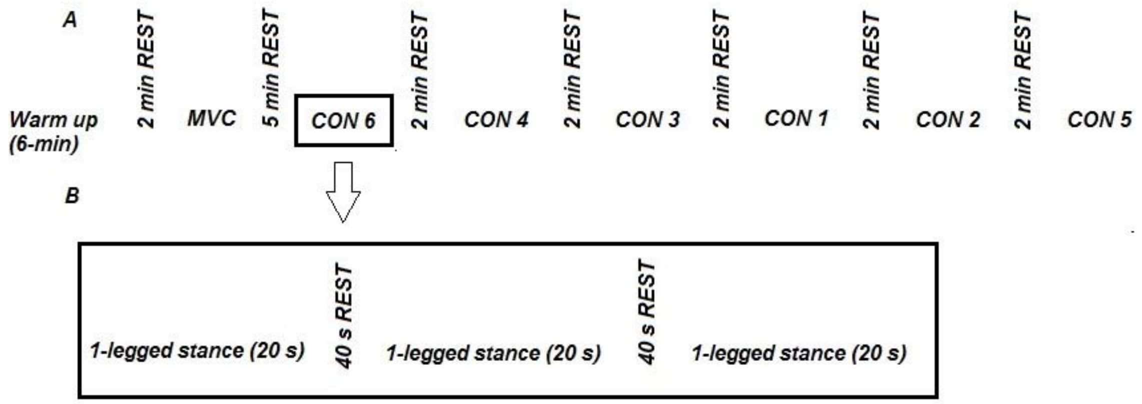

Figure 1

Overview of the experimental protocol.
Panel A represents timeline of the experimental protocol. Panel B represents timeline of one leg stance in all 6 conditions (CON). Conditions (stability levels) were randomized. MVC – maximal isometric plantar and dorsal flexion measurements.

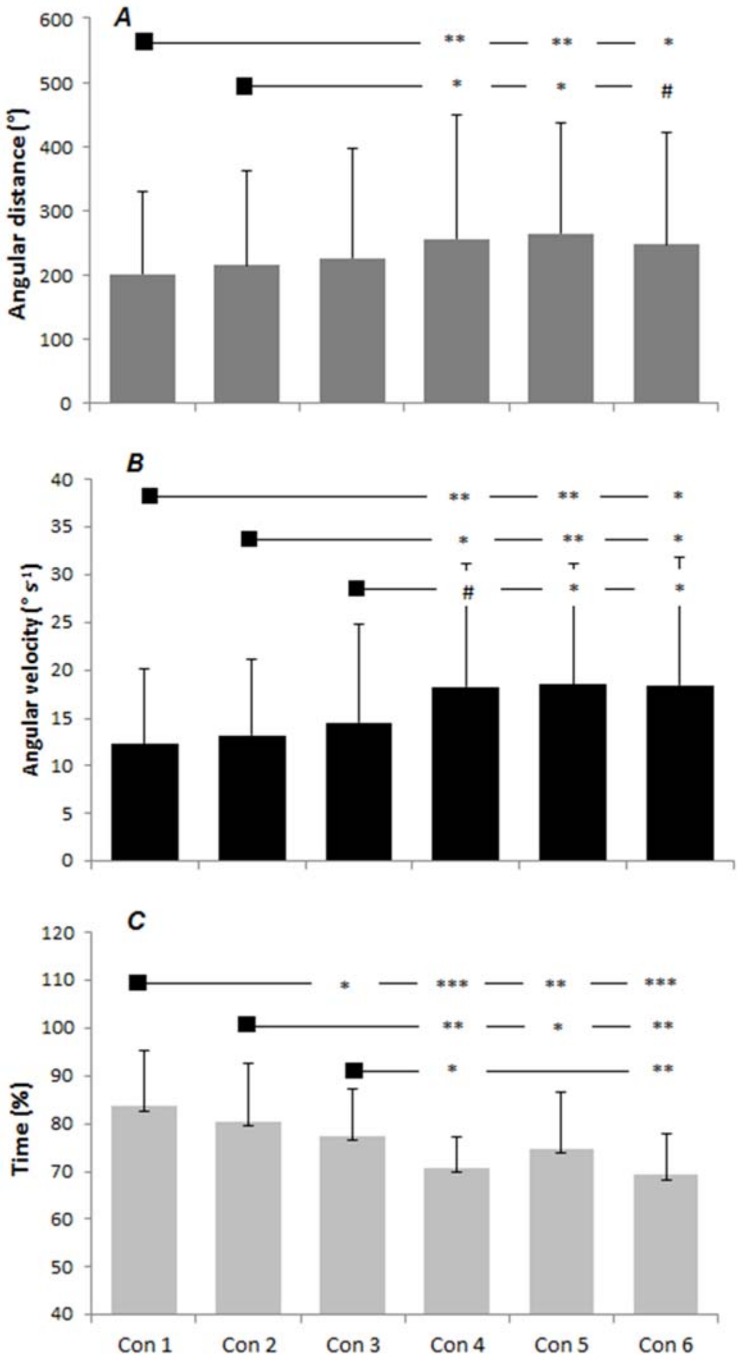

Figure 2

*Comparison of exercise difficulties between conditions. Panels A and B represent angular distance (degrees) and velocity (degrees ×s-1) of T-board swaying. Panel C represents the percentage of a total balance time in one repetition (%). Asterisks indicate significant differences between the condition identified with a square and level corresponding to the asterisks (*p < 0.05; **p < 0.01; ***p < 0.001). Section sign indicates differences on the verge of significance between the condition identified with a square and level corresponding to the section sign (p < 0.09). Con – condition.*

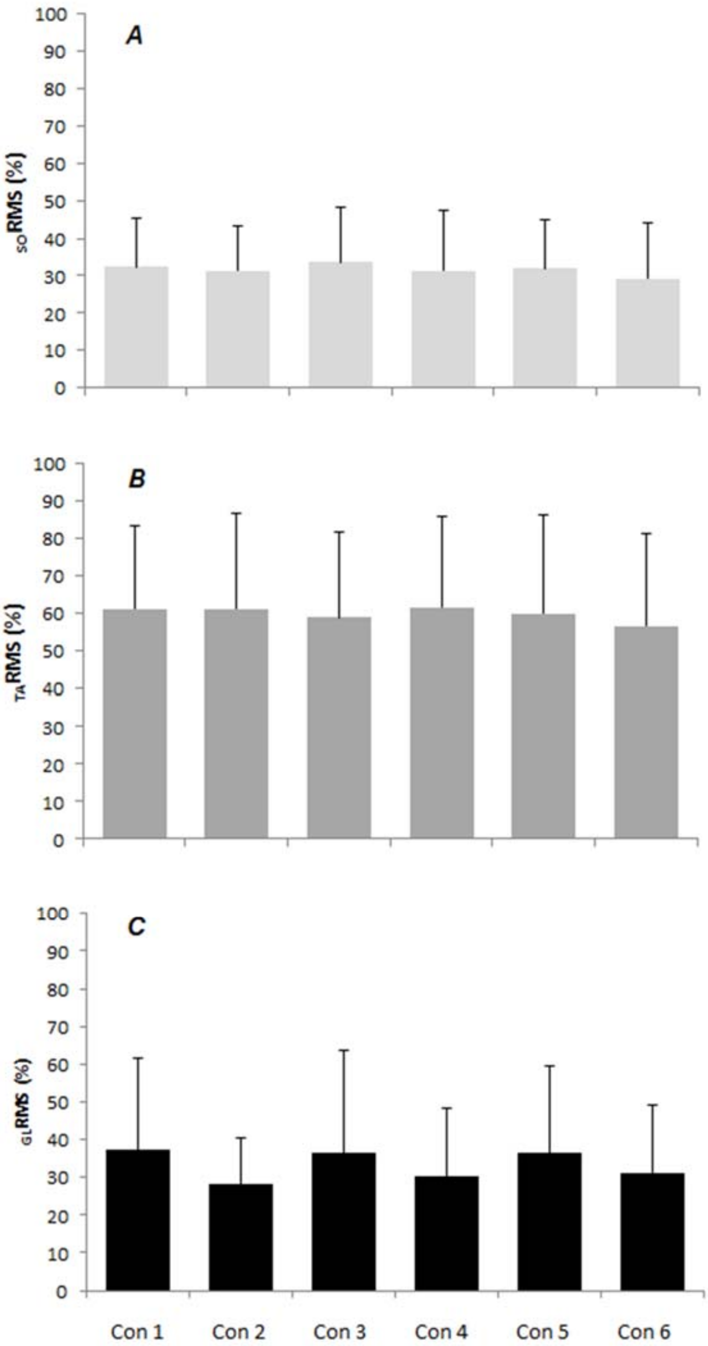

Figure 3

Comparison of ankle muscle activation during a one leg stance on a T-board. RMS – the root-mean-square of the surface electromyographic signal (EMG), SO – soleus, TA – tibialis anterior, GL – lateral gastrocnemius, Con – condition. Data are expressed as the mean percentage of maximal surface EMG during maximal plantar and dorsal flexion.

Table 1
Participants' averages of root-mean square values of surface EMG during increased difficulties of a one-*leg stance.*

One leg stance	$_{SO}$RMS (μV)	$_{TA}$RMS (μV)	$_{GL}$RMS (μV)
Con 1	76.3 (39.6)	277.7 (120.6)	57.8 (34.9)
Con 2	72.3 (32.9)	270.2 (105.7)	46.6 (32.7)
Con 3	80.9 (48.0)	259.3 (94.2)	57.6 (46.4)
Con 4	71.2 (37.1)	271.6 (96.0)	47.6 (32.5)
Con 5	75.7 (40.3)	271.7 (151.7)	60.8 (49.2)
Con 6	66.5 (34.4)	244.5 (87.2)	50.3 (37.3)

Data are expressed as means ± standard deviations. $_{SO}$RMS – averages of root mean square values of the solelus ($_{SO}$RMS), tibialis aterior ($_{TA}$RMS) and lateral gastrocnemius ($_{GL}$RMS) surface EMG signal. Con – condition.

Discussion

The objective of this study was to investigate ankle muscle activity (i.e., m. soleus, m. tibialis anterior, m. lateral gastrocnemius) during a one-leg stance on a T-board with a graded level of instability in the sagittal plane. It was hypothesised that an increment in difficulty of the one leg stance on a T-board would be coupled with an increase in the percentage of maximal ankle muscle activity and could be used as an outcome of relative measure of intensity of balance exercise. The present results indicate that graded levels of T-board swaying did not generate significantly higher ankle muscle activity. In conclusion, the percentage of ankle maximal muscle activity could not be used as a relative measure of intensity for a balance exercise.

In the current study, an increment in the difficulty of the one leg stance on a T-board, i.e., increment in angular velocity of the T-board, resulted in a decrease in active balancing time (Figure 2). Specifically, it was easier for participants to maintain equilibrium; i.e., both ends of the boards raised from the floor, under the first three conditions (Con1-3) compared to the last three ones (Con4-6; Figure 2C), thus the percentage of active balancing time decreased from ~84% under Con1 to ~69% under Con6. Significantly greater (i.e., ~201° vs. ~250°) and faster (i.e., ~12°s-1 vs. ~18 °s-1) swaying of the support surface probably triggered more perturbation, which could not be so easily counterbalanced by corrective postural adjustments, therefore, reduction in active balance time was obtained. It should be emphasized that the significant differences in active balance time were not obtained between all levels, nevertheless, there were significant differences between the first three and the last three conditions (Figure 2). Accordingly, two grades of balance exercise difficulties can be recognized. Irrespective of the two difficulty levels of the one-leg stance, the percentage of maximal ankle muscle activity (Figure 2) did not change significantly with an increase in T-board instability. Thus, the expected increment in muscle ankle activity due to greater angular velocity of the T-board could not be proven (Figure 3).

Likewise, Laudner and Koschnitzky (2010) did not obtain significantly greater increases in ankle muscle activation when comparing a single-leg stance on a BOSU balance trainer with the base (hard flat side) down or with the base up (inflatable bladder) on the ground. Moreover, Borreani et al. (2014) showed that tibialis anterior and soleus activation did not increase between one leg stances on a stable surface compared to a rocker board that was unstable in the sagittal plane. In this study they also found that a multiaxial platform did not provoke significantly greater ankle muscle activation than a uniaxial platform when the same one leg stance was used (Borreani et al., 2014). On the other hand, Muehlbauer et al. (2014) showed that increased sensory difficulty in a one-leg stance resulted in deterioration in balance performance and increases in muscle activity for all (i.e., soleus, tibialis anterior) but two muscles (i.e., gastrocnemius, peroneus). Any discrepancies in obtained ankle muscle activation data between studies could be related to the different movement strategies used by participants when recovering stability in response to the displacement of the supporting surfaces (e.g., balance board, foam, T-board). The motor patterns used for controlling stable posture and stance have been described as ankle, hip, step and reach-to-grasp strategies (Shumway-Cook and Woollacott, 2017). The ankle strategy relies on relatively stereotyped patterns of leg and trunk muscle activation, which begin in the ankle joint muscles and then radiate in sequence to thigh and trunk muscles, while the hip strategy relies on relatively stereotyped patterns of leg and trunk muscle activation, which begin in the hip and trunk muscles, whereas the ankle muscles are generally unresponsive (Horak and Nashner, 1986; Woollacott and Shumway-Cook, 1999). It might be possible that a hip strategy or mixed strategy, i.e., ankle plus hip, were used to a greater extent than a pure ankle strategy to restore equilibrium on the BOSU balance trainer (Laudner and Koschnitzky, 2010), multiaxial platforms (Borreani et al., 2014) and the T-board used in our study. Thus, activation of muscles at the ankle did not change significantly with increased compliance and/or instability of support surfaces, i.e., multiaxial platforms, BOSU balance trainer with the base (hard flat side) down and higher angular velocity of the T-board. Additionally, Horak and Nashner (1986) suggested that a hip strategy was used to restore balance in response to larger and faster perturbations or when the support surface was compliant.

Thus, the increment of angular velocity of T-board swaying increased postural demands in a one-leg stance to such extent that stability could not be compensated solely by increasing ankle muscle activity. Although ankle strategy is more effective, it is likely to cause a greater error in the centre of mass displacement than hip strategy (Woollacott et al., 1988), thus, leading to a greater possibility of T-board edges touching the ground. Subsequently, the hip strategy was likely superior in regaining equilibrium during a one-leg stance when participants were exposed to faster T-board swaying. The abovementioned explanation could be further supported by the Mergner's model of vestibular-somatosensory interaction, which states that healthy subjects use vestibular information for postural orientation during standing only when the support surface is interpreted as unstable (Mergner and Rosemeier, 1998). Although not measured, it is possible that greater engagement of the vestibular system due to faster swaying of the T-board in the sagittal plane provoked mostly proximal (e.g., hip and trunk muscles) rather than distal (e.g., ankle) sequences of muscle activation to successfully re-establish equilibrium during the one-leg stance for the last three conditions (Con4-6). Therefore, the percentage of maximal muscle activation solely at the ankle could not be used as a relative measure of balance exercise intensity. On the other hand, the percentage of active balance time might be a way to differentiate between two difficulties (i.e., below 70% and above 70% of a total balance time).

Taking all results into consideration, the findings of the current study imply that the percentage of maximal ankle muscle activation is not a good outcome variable to describe a change in 'intensity' in a one-leg stance on a T-board. Thus, monitoring other corporal segments, i.e. trunk and hip muscles to obtain their responses to incrementally increasing levels of balance exercise difficulty, e.g. active balancing time, would likely reveal different results.

The current study does have some limitations. One limitation is related to the limited number of muscles monitored. Hip and trunk muscular activity measurement during balancing on the T-board could reveal additional insight into variations of balance strategies. Moreover, an increase in facial muscle activation has been demonstrated in strenuous lower limb exercise (de Morree and Marcora, 2010), thus, their activation level might also add significant insight into individual responses to different levels of difficulty in balance exercise. The second limitation of this study is related to a non-linear increase in one leg stance difficulties, i.e., angular velocity. There was an underlying assumption that all individuals recruited in the present study had similar maximal balance capacity and were consistently challenged to the same extent at the next level of T-board swaying. We tried to overcome the aforementioned limitation by ensuring that all participants were healthy adults, with no history of ankle injuries and recreationally trained, thus the chosen balance exercise was probably performed near the same limits of maximal balance capacity. Furthermore, balance emerges from the interaction of different postural sub-systems, e.g., musculo-skeletal, muscle synergies, sensory systems, sensory organization, cognitive strategies and resources (Shumway-Cook and Woollacott, 2017). Despite the fact that muscle activity ensures the generation of sufficient forces to produce movements in controlling the body's position in space, muscle activity is just one aspect of one single component of balance. In conclusion, there is still an important gap in the development of an objective instrument that can report the intensity level of balance exercise. Postural control requires highly complex interactions of the human neuromuscular system, so future studies should focus more on postural synergies, i.e., a combination of control signals sent to several muscles to ensure the stability of a whole body (Latash, 2008) in response to increases in difficulty of the balance exercise. Moreover, balance training is also frequently implemented with the intention to prevent ankle sprains. Thus, a plateau in difficulty of balance exercise for enhancement of ankle muscle activity can also be suggested.

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