Does dual tasking affect the ability to generate anticipatory postural adjustments?

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List of Abbreviations

BF	Bicep Femoris
CB	Counting Backwards
CNS	Central Nervous System
COM	Centre of Mass
COP	Centre of Pressure
EMG	Electromyography
GM	Gastrocnemius Medialis
NCT	No Cognitive Task
RF	Rectus Femoris
SD	Standard Deviation
SE	Standard Error
ST	Stroop Task
TA	Tibialis Anterior

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Declaration

I declare that the work submitted in this thesis is my own and has not been submitted as part of any other degree or qualification. Studies described in this thesis have been acknowledged and cited accordingly.

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Abstract

Introduction: To date, little is known about the impact of additional cognitive tasks on perturbed balance and whether different types of cognitive tasks can generate different balance mechanisms. The aim of the study was to investigate how two different cognitive tasks (Stroop test and counting backwards task) would influence young adults' ability to generate appropriate postural responses. **Methods:** Twenty young adults (25.95 ± 2.97 years) were asked to stand eyes open, bare feet shoulder-width apart on a moving platform which was translated in the anterior-posterior direction at three different frequencies (0.1, 0.25, 0.5 Hz) and perform either a counting backwards task, a Stroop task, or no cognitive task. Tonic activity and muscle onset latencies of the Rectus Femoris, Bicep Femoris, Tibialis Anterior and Gastrocnemius Medialis muscles were measured through surface electromyography (1000 Hz), and the number of cognitive errors was recorded. **Results:** Results showed no significant differences in muscle onset latencies and tonic activity between dual tasking and single tasking conditions, nor between the two dual tasking conditions. More cognitive errors were made in the counting backwards task (238 total cognitive errors across all frequencies) compared to the Stroop task where no errors were recorded. A frequency effect was identified with participants, regardless of condition, showing greater tonic activity in the Rectus Femoris (p= 0.012, M= 177% baseline, SD= 79.2), the Gastrocnemius Medialis (p= 0.016, M= 274.8% baseline, SD= 201.4) and the Bicep Femoris (p= 0.043, M= 291%, SD= 3.5) at 0.5 Hz, as well as earlier muscle activation in the Tibialis Anterior (p< 0.001, M= -2.7, SD= 8.1% half cycle), the Gastrocnemius Medialis (p< 0.001, M= -9.54, SD= 3.3% half cycle) and the Bicep Femoris (p< 0.001, M= -1.34, SD= 3.9% half cycle) at 0.5 Hz compared to the other frequencies. Transition and steady state muscle onset latencies were only significantly different for the Gastrocnemius Medialis at 0.25 Hz (p= 0.001), possibly because the 0.1 Hz frequency was too easy to require adaptation and the 0.5 Hz frequency was large enough to trigger earlier muscle activation from transition state which was then carried to steady state. Dual tasking did not seem to influence anticipatory postural adjustments, however perturbation intensities did. **Discussion:** It is assumed that due to the 'threatening' nature of the 0.5 Hz perturbation, a stiffer position was adopted as seen by the increased tonic activity, and anticipatory mechanisms were triggered sooner than the other frequencies, as seen by

earlier muscle activation. Since posture was unchanged between single and dual tasking, it is suggested that participants' postural control was automated and the cognitive errors in the two mental tasks could reflect their difficulty level. Future research should explore body kinematics to identify the balance strategies adopted, as well as take into account the reaction time of the cognitive task to better understand participants' allocation of attention during perturbed balance dual tasking.

1. Introduction

1.1 Balance mechanisms

Balance is commonly defined as the ability to keep the body's center of gravity within its base of support and can be characterized as either static or dynamic balance (Goldie et al., 1989). Static balance refers to one's ability to maintain their centre of gravitywithin their base of support while maintaining an upright position during quiet stance. Conversely, dynamic balance refers to the maintenance of an upright posture while one's centre of gravitymoves outside their base of support (e.g., walking, running, etc.) (Yim-Chiplis and Talbot, 2000). While testing static balance can provide insight into passive postural control and the effects of small, self-initiated corrective movements (Visser et al., 2008), investigating dynamic balance can provide insight into the performance of athletic activities, as well as daily living, due to its functional nature. For example, one of the most common dynamic activities of everyday life is walking (Hamacher et al., 2011) which for many older adults, including those with pathologies such as Parkinson's disease (Nantel et al., 2011), carries a risk of falls (Fletcher and Hirdes, 2002). As such, investigating dynamic balance is of immense importance.

1.1.1 Systems contributions to balance

For the achievement, maintenance, or regaining of balance, the combination of the visual, vestibular, and somatosensory systems is essential (Gaerlan et al., 2012). Visual input allows the central nervous system (CNS) to use external, static references (e.g., wall) to determine a vertical frame of reference and interpret movement of the body relative to this point (Merla and Spaulding, 1997). Vision is the most important sensory system used in balance (Uchiyama and Demura, 2009), and is comprised of three components: central (or focal), ambient (or peripheral), and retinal slip. The central visual system contributes to object recognition and object motion perception, while the ambient visual system contributes to spatial orientation and postural control in locomotion. The retinal slip refers to the image of the visual surroundings moving on the retina and can be caused by either horizontal or vertical movements of the head while visually fixating on an object, or target (Herdman et al., 1995).

The importance of vision has been widely investigated using two paradigms: i) a 'sensory restriction' and ii) a 'sensory conflict' paradigm. In the first paradigm, sensory restriction, balance performance is observed while visual information is not available. Research suggests that postural stability is significantly reduced, as expressed by increased sway area, path length and centre of pressure (COP) velocity, when adults, as well as children, are asked to stand in a dark environment, compared to standing in a lit environment (Ashmead and McCarty, 1991, Rougelj et al., 2014). The second paradigm, sensory conflict, involves maintaining balance in situations where the visual information available contradicts information from the other sensory systems. A very common experimental condition used to test this is that of the 'moving room', where the walls and ceiling move forward and backward while the floor is stationary (Mahboobin et al., 2005, Sparto et al., 2006). When faced with discrete movements of their visual surrounds, older adults swayed more than young adults (Wade et al., 1995), however, Prioli et al. (2005) identified that only sedentary older adults displayed increased body displacement compared to active older adults and young adults. These results suggest that physical activity can improve older adults' ability to deal with visual manipulations and more accurately integrate visual information with vestibular and somatosensory information to generate the correct motor action.

The vestibular input is an internal reference that measures the orientation of the head in space and facilitates the stabilization of the eyes, contributing to the maintenance of balance during quiet stance, as well as during walking (Shumway-Cook and Woollacott, 2007). The main components of the vestibular system are found in the vestibular labyrinth, which is made up of the semicircular canals and the otolith organs in the inner ear (Khan and Chang, 2013). The semicircular canals are highly sensitive to rotational movements (Rabbitt, 1999), while the otolith organs detect linear acceleration (Grant and Best, 1987). The brain then combines both rotational and linear acceleration into a resolution of motion and orientation relative to the environment (Raphan et al., 1996). When it comes to identifying self-generated and externally-generated movements, the brain makes internal predictions of sensory input based on the proposed actions, which are then contrasted against the actual sensory input of the movements. The difference between the predicted and actual sensory input of the movement is perceived as self-generated movement (Angelaki and Cullen, 2008).

The somatosensory input offers information regarding the position of the body by using proprioceptors (e.g., muscle spindles, Golgi tendon organs, and joint receptors) (Shaffer and Harrison, 2007)) found in the muscles, tendons, and joints. Such information include motion and loading of the muscles and joints, and once processed by the brain, corrective mechanisms of the musculoskeletal system are generated, contributing to the maintenance of balance (Cnyrim et al., 2009). For instance, during body sway, a *passive* torque is created around the ankle joint due to the acceleration of the body created by gravity. Starting from the joints and muscles, this information travels to the brain as sensory feedback, where a corrective, *active* torque is applied in order to maintain balance (Peterka, 2000).

The visual, somatosensory and vestibular systems provide information associated to changes in body orientation (due to self-motion or external perturbation) which are then integrated in the CNS. One of the proposed mechanisms by which the CNS processes multisensory information is the "weight and re-weight" mechanism (Horak, 2006). After the information from each individual sensory system is combined, a weight is assigned to each input source based on the current functional state of the sensory systems as well as the nature of the postural task and the circumstances it is performed under (Kabbaligere et al., 2017). As the sensory environment is changed, the dependence on each sensory input needs to be re-weighed (Horak, 2006). For instance, healthy adults have a 70% dependency on somatosensory input, 10% dependency on visual input and 20% dependency on vestibular input when standing in a well-lit environment and with a firm base of support (Peterka, 2002). However, when standing on an unstable surface, dependency on the vestibular and visual system is increased, as dependence on surface somatosensory inputs for postural orientation is decreased (Peterka, 2002). Re-weighing is, therefore, important for maintaining balance under various environmental conditions. When one of the neurophysiological systems is damaged, the ability to quickly re-weigh sensory dependence becomes impaired, possibly because the neural systems are linked and overlap (Mast et al., 2006), while when more than one system is damaged, one's ability to maintain balance is negatively affected (Ray et al., 2008).

1.1.2 Synergies

During maintenance of postural control, it is crucial for the CNS to invoke postural control mechanisms rapidly. This can be achieved with the use of muscle synergies. Torres-Oviedo and Ting (2007) define muscle synergy as a group of muscles that are activated synchronously and have fixed relative gains. By using muscle synergies, there is less demand on the CNS as a single neural command can recruit a set of muscles (Torres-Oviedo et al., 2006). It has been suggested that there is a limited number of synergies and, in the example of frogs, certain tasks (e.g., walking or jumping) require the use of specific synergies, while there are also synergies that are shared across the same tasks (d'Avella and Bizzi, 2005). Muscle synergies have been observed in many contexts of human locomotion, including during quiet standing and perturbed stance. In quiet standing, synergies have been found to coactivate muscles throughout the limbs and trunk (Ting and Mckay, 2007) while in regard to perturbations, one or more muscle synergies can be activated and the combination of these will influence the subsequent muscle activation patterns (Ting, 2007). In young adults, muscle synergies can be activated about 200-300 ms prior to the initiation of a platform perturbation (Piscitelli et al., 2017). This feedforward mechanism is referred to as anticipatory synergy adjustments and its purpose is to facilitate the stabilisation of the coordinates of the COP (Klous et al., 2011).

1.1.3 Postural control strategies

More often than not, during normal stance we are faced with externally induced perturbations that challenge our balance (e.g., standing on a moving bus). On such occasions, postural muscles need to be activated in order to restore the center of mass (COM) stability. It has been found that the CNS typically uses three strategies: two feet-in-place strategies -- the ankle and the hip strategies -- and a stepping strategy, which can be used either separately or combined to restore balance (Nashner and McCollum, 1985). When the perturbation takes place in the anterior-posterior plane, an ankle or hip strategy is most commonly used (Horak and Nashner, 1986), while a hip or stepping strategy is likely to be adopted when the perturbation lies in the medial-lateral plane or the anterior-posterior perturbation is too large, (Winter et al., 1996). The speed of the perturbation also affects the choice of the response used for the recovery of balance. Hwang et al. (2009) reported that during slow-speed (0.1 $m \cdot s^{-1}$) anterior-posterior perturbations, young adults used an ankle strategy, while during fast-speed (0.2 $m \cdot s^{-1}$) anterior-posterior perturbations a mixed ankle-hip strategy was used.

When the ankle strategy is used, the body acts as a single-segment inverted pendulum and torque is produced around the ankle. In response to posterior platform translations, healthy adults sway their body forward and muscle activation is first observed in the gastrocnemius, then hamstrings and erector spinae (Nashner, 1976). During quiet standing, the ankle strategy is thought to be advantageous over the other strategies as upright stance is maintained with minimal head movements, suggesting that vestibular and visual feedback is improved (Assaiante and Amblard 1995; Kuo 1995).

When ankle torque is ineffective at producing whole-body motion (i.e., when the support surface is compliant or too small, such as when balancing on a beam, or when the platform perturbation speed is too large), the hip strategy is employed acting as a double-segment inverted pendulum where the COM is stabilised by rotating the upper body forward and downward (about the pelvis/hip joints) and rotating the lower body backward (about the ankles) (Horak and Nashner, 1986). During posterior platform translations, muscle activation is first observed in the abdominal muscles followed by the quadricep muscles (Shumway-Cook and Woollacott, 2007), while during medial-lateral platform translations, movement is detected at the pelvis, where adduction of one leg and abduction of the other leg is required (Winter et al., 1996). Kuo and Zajac (1993) found that when using the hip strategy, the capability to accelerate the COM without taking a step is increased and the immediate muscle activity required to accelerate the COM is lower than when an ankle strategy is used, hence why a hip strategy is used when dealing with more challenging conditions. It is worth mentioning however, that they only investigated the instantaneous muscle activity, therefore it is unclear whether the total muscle effort is smaller for the hip than for the ankle strategy.

In situations where the perturbation is too large and a feet-in-place response is not effective, falling is avoided by taking a step (Burtner et al., 2007; Roncesvalles et al., 2000). The perturbation conditions are not, however, the only factors that affect the selection of a postural response. Factors such as experience and adaptation (Welch and Ting, 2014), as well as fear of falling (Adkin et al., 2000) also play an important role. For instance, Adkin et al. (2000) investigated the effects of increased postural threat by increasing platform height in young adults and observed that a tighter postural control (as seen by decreased COP displacement) was adopted and that this response was scaled to the degree of postural threat, suggesting that psychological factors, such as fear of falling, can also impact postural responses.

1.1.4 Anticipatory and compensatory postural control

There are generally two ways of controlling movement that dictate the strategies used. For instance, when faced with a discrete unexpected perturbation, such as a single movement of the support surface resembling a trip or slip (Horak et al., 1997), humans use compensatory postural adjustments (i.e., activation of postural muscles *after* the perturbation has occurred) to correct for the shift in COM (Welch and Ting, 2008). However, when a perturbation is predictable, anticipatory postural adjustments are made by activating postural muscles *in advance of* the upcoming disturbance, therefore reducing the need for large compensatory postural adjustments after the perturbation (Frank and Earl, 1990; Pavol and Pai, 2002). Anticipatory postural adjustments can also be elicited during whole-body movements, such as the initiation of gait. During gait initiation for example, decreased activity of the soleus muscle combined with an activation of the tibialis anterior muscle (Crenna and Frigo, 1991) causes an anticipatory COP shift toward the swing leg (i.e., forward), establishing an effective contact position for the swing foot and a stable body progression (Honeine et al., 2016). Even though both anticipatory and compensatory mechanisms improve gradually until adulthood, compensatory processes are controlled much earlier in development than anticipatory processes (Hay and Redon, 1999). For example, to investigate the contribution of anticipatory and compensatory mechanisms during development, Hay and Redon (1999) asked children (3-10 years old) and adults to perform an unloading task. The task involved participants holding a load in their hands, with arms by their side and forearms horizontal in front of them, while standing eyes-closed and the load was either unpredictably removed by the experimenters, or voluntarily removed by the participants. Their results suggested that although both groups used anticipatory mechanisms to control their posture in preparation for the unloading disturbance, it was revealed that the adult group was more successful in using anticipatory mechanisms than the young group. Palluel et al. (2008) found that "adult-like" anticipatory strategies fully develop around the age of 12 after testing children between the ages of 8 to 12 using a leg raising task.

While discrete movements and perturbations can elicit either a compensatory or an anticipatory mechanism, continuous perturbations, such as those experienced while standing on a bus, can evoke both responses, offering a more robust understanding of the mechanisms of postural control (Schmid et al., 2011). In a laboratory setting, the oscillating platform paradigm, where the support surface is perturbed at various frequencies and amplitudes, provides insight into the ability to switch between anticipatory and compensatory mechanisms. A compensatory response is stimulated by the initial perturbation and as the platform continues to oscillate, a switch to an anticipatory mechanism is observed (Schmid et al., 2011). For instance, Bugnariu and Sveistrup (2006) found that younger adults shifted from compensatory postural adjustments to anticipatory postural mechanisms (evidenced through earlier postural muscle onsets) within the first three to five cycles of externally induced sinusoidal platform oscillation, while older adults were unable to make such shifts. Prior experience of the perturbation has also been shown to affect anticipatory postural adjustments (Kennedy et al., 2013; van Ooteghem et al., 2009; Nashner, 1976). Kennedy et al. (2013) for instance, found that young adults carried the experience gained from one postural trial to the following trials and used this experience to decrease their COP displacement and to activate their postural muscles earlier in anticipation of the perturbation. They also observed that after seven oscillation trials (lasting 1 minute each), further experience did not have any additional effect on anticipatory postural responses.

1.2 Balance and dual tasking

Circumstances requiring the processing of motor and cognitive tasks simultaneously (i.e., dual tasking) constitute a significant component of the modern busy lifestyle (e.g., talking on the phone while standing or walking). As such, it is important to consider the interaction between cognitive processing and motor performance. Working memory, which has limited capacity and consists of three components: central executive (responsible for the allocation of attention to information in the environment); phonological loop (involved in verbal rehearsal); and visuospatial sketchpad (involved in visual and spatial processing and storage), temporarily holds information necessary for cognitive processing (Baddeley and Hitch 1974). When a task is performed, a portion of this limited information processing capacity is required (Kahneman, 1973). Consequently, during dual tasking, the performance of one, or both tasks can decrease if the task requirements exceed the available capacity.

According to the Cross-Domain Competition Model (Lacour et al., 2008), since postural and cognitive tasks compete for attentional resources, postural performance during dual tasking should be inferior to the single postural task performance. Several studies have shown that balance performance is compromised when cognitive and motor tasks are performed simultaneously (Andersson et al., 1998; Pellecchia, 2003b; Mitra and Fraizer, 2004). For instance, Pellecchia (2003b) found that during dual tasking, where participants performed three information reduction tasks while stood on a compliant surface, balance performance decreased (i.e., postural sway increased) in both young and older adults. Furthermore, postural sway increased linearly with the difficulty of the cognitive task. Reduced balance performance during dual tasks is mostly observed in older populations (such as, changes in the COP and/or larger sway area), whether a quiet stance is adopted (Maylor and Wing, 1996) or under a more demanding postural task (i.e., standing on a sway-referred platform) (Andersson et al., 1998) due to decreased cognitive and attentional capacities (Shumway-Cook and Woollacott, 2000). Other studies, however, have observed contrasting results in young (Riley et al., 2003; Swan et al., 2004) as well as older populations (Deviterne et al., 2005). Riley et al. (2003) found that postural sway was reduced when young adults performed a digit rehearsal task while stood on a foam surface. Beretta et al. (2019) observed that young adults reduced their COP sway during dual tasking on a moving platform, while Dault et al. (2001) found that young adults improved postural stability, as seen by their increased frequency and decreased amplitude of sway, when working memory tasks of different difficulty level were added. Deviterne et al. (2005) reported that when listening to a meaningful auditory message (i.e., a story), older participants displayed better balance performance values, as seen by their reduced sway area and sway path. To ensure that participants would focus their attention on the story, the experimenter would ask them questions about the story they had just listened to at the end of the test.

The mixture of results found in dual task literature suggests that the Cross-Domain Competition Model fails to explain why, in some cases, postural stability has been enhanced. It has been suggested that these differences can be a result of the various difficulty levels of postural tasks used in the literature (e.g., quiet stance, tandem stance vs perturbed stance), as well as the difficulty of the cognitive tasks (Wollacott and Shumway-Cook, 2002). For this reason, a U-shaped nonlinear interaction model was proposed (Huxhold et al., 2006), suggesting that the cognitive demand of the secondary cognitive task (low or high), can either improve or diminish balance performance. Huxhold et al. (2006) used three cognitive tasks of different difficulty levels during quiet stance and observed that both young and old participants reduced their COP, therefore improved balance performance, when performing the easy cognitive task (choice-reaction time task). However, when the demand of the cognitive task increased, young participants exhibited unchanged body sway, while older participants increased their body sway. Deviterne et al. (2005) showed improved balance when older participants attended to an auditory task. This could be due to the auditory task being a very low-demanding secondary task, hence why balance performance was improved. Another possible reason for these results could be associated with their shift of attention towards the cognitive task and away from the postural task, leading to a more automatic processing of posture (Wulf et al., 2004).

The type of cognitive task performed during dual tasking can, therefore, also influence balance performance. Maylor et al. (2001) found that compared to the spatial Brooks' task (participants were instructed to place consecutive numbers in a 4x4 grid, e.g., `In the next square to the right/left/up/down put a 2'), both young and older adults had higher sway velocity when performing the non-spatial Brooks' task during quiet stance. The non-spatial task was based on the stimuli from the spatial task, replacing the words right, left, up, and down, with the words quick, slow, good, and bad, respectively. Jamet et al. (2004) also showed that during quiet stance, a counting backwards mental task induced postural sway in older adults, while the Stroop test did not. For the correct execution of visuo-verbal tasks, like the Stroop test, accurate visual fixation and focused attention on the coloured word is necessary (MacLeod, 1991). Therefore, the authors (Jamet et al., 2004) argued that visual landmark usage can compensate for the adverse effects of added cognitive load on balance, while on the other hand, counting backwards does not require gaze fixation for its execution and environmental information is not taken into account, destabilising postural control.

Dual tasking can also affect the balance strategy used. For example, the use of the stepping strategy is elicited following discrete perturbations combined with a secondary cognitive task (i.e., counting backwards by 3) (Rankin et al., 2000). The authors found that due to the increased attentional demands of dual tasking, the feet-in-place strategy was less effective for balance recovery and therefore participants took more steps. This was seen both in young and older adults, however, older adults stepped more frequently in comparison to their young counterparts, possibly due to age related deficits in the allocation of attention (Weeks et al., 2003). In regard to continuous perturbations, Laessoe and Voigt (2008) found that only older adults increased their stepping frequency during dual tasking when young and older adults were subjected to predictable perturbations. It can be argued that the predictability of the perturbation could have influenced young adult's stepping responses. Indeed, when young adults were in control of the perturbations (self-triggered), Bugnariu and Sveistrup (2006) found that no steps were taken, compared to externally- triggered perturbations that elicited stepping reactions.

Research has shown that when a cognitive task is performed during quiet stance, balance performance is affected, however, the results are conflicting, with some studies showing a decrease in performance (Mitra and Fraizer, 2004) and others showing an improvement (Riley et al., 2003). Since cognitive and postural tasks share the same resources (Woollacott, 2000), the difficulty level of the cognitive, or motor task (Huxhold et al., 2006), the focus of attention (Wulf et al., 2004), as well as the different types of cognitive tasks performed during quiet stance (Maylor et al., 2001; Jamet et al., 2004) can affect postural control.

1.3 Aims and hypothesis

To date, little is known about the impact of cognitive tasks on perturbed balance and whether different types of cognitive tasks elicit different balance mechanisms.

The aim of the study was therefore to investigate how dual tasking influences young adults' ability to generate anticipatory postural adjustments. It was hypothesised that participants would i) take more steps, make more cognitive errors, and display delayed postural muscle onset latencies as well as greater tonic activity during dual tasking conditions compared to the single task condition; ii) not be able to shift from reactive to anticipatory postural mechanisms (as evidenced by the timing of the activations) during dual tasking when compared to single tasking; and iii) display different muscle onset latencies and tonic activity between two cognitive task conditions.

2. Methods

2.1 Participants

Twenty young adults (11 females and 9 males) between the ages of 21-31 years were recruited on a voluntary basis through word of mouth. Mean age (±SD) was 25.95 (± 2.97) years, mean height (±SD) was 172.69 (±8.75) cm and mean mass (±SD) was 70 (±14.1) kg. All participants were free from any neuromuscular disorder and had no existing or unresolved injuries that could limit movement in any way. The study was reviewed and approved by Manchester Metropolitan University's ethics committee and written consent from each participant was obtained before taking part in the experiment.

2.2 Procedure

Participants attended the laboratory at Manchester Metropolitan University for one 90-minute session. They were asked to stand eyes open, bare feet shoulder-width apart on a moving platform which was translated 20 cm peak-to-peak in the anterior-posterior direction at three different frequencies (0.1, 0.25, 0.5 Hz). Trials for each frequency were 100s long and consisted of at least 10 cycles at 0.1 Hz, 20 cycles at 0.25 Hz and 40 cycles at 0.5 Hz (Mills and Sveistrup, 2018). Participants were tested under three conditions: a) Stroop Test (ST), b) Counting Backwards (CB) and c) No Cognitive Task (NCT) and each condition was performed while stood on the moving platform in a well-lit laboratory environment. For the ST, participants were presented with several coloured words, representing colour names that are different from the printed colours. They were instructed to name the colours of the words as quickly as possible. For example, if the word was "yellow" and it was printed in red ink, the correct answer would be "red". The ST was performed using PsychoPy software (Peirce et al., 2019). Words appeared on the screen one at a time every four (4) seconds. Figure 1 shows a participant while performing the ST. For the CB condition, they were given a random number over 100, from which they had to start counting backwards by seven (Maclean et al., 2017), as fast and as accurately as possible for the duration of the trial. Numbers over 100 were chosen to ensure participants would not count below zero. An audio recording device was used to record their answers so that the number of errors for both cognitive tasks could be calculated. For the NCT, participants were instructed to focus on a cross projected on a screen in front of them while stood as still as possible. In all conditions, if/when a step was taken, participants were instructed to regain their balance and return to their initial position. The number of steps taken by each subject at each frequency was documented. Participants also performed the CB and ST while stood in quiet stance on the platform. Quiet stance trials were 30s long and were recorded in order to compare dual tasking cognitive performance (i.e., number of errors made) to single tasking. Two trials of each condition in randomised order were performed. Participants were equipped with a harness, which was attached to an auto belay from the ceiling, in case the perturbation caused a fall.

Figure 1. Participant performing the Stroop Task while stood on the moving platform.

2.3 Electromyography

Postural muscle activation was recorded via surface electromyography (EMG) (Delsys Trigno, Delsys Inc, USA) (1000 Hz). Surface electrodes were placed on the skin over the muscle belly of the tibialis anterior (TA), gastrocnemius medilalis (GM), rectus femoris (RF) and bicep femoris (BF) on the left side of the body following the SENIAM guidelines (Hermens et al., 1999). Shaving of hair and cleansing these specific areas with alcohol wipes was necessary to remove dead surface tissue and oil that could potentially reduce conductivity. To ensure appropriate EMG signal, its quality was checked visually at the time of the recording.

2.4 Data Analysis

2.4.1 Postural muscle onset latency and tonic activity

Data were processed offline. Motion capture data were individually reconstructed and digitally labelled in VICON Nexus 2.11; marker trajectories and joint kinematics[1] were then exported for data analysis. Data were processed using Microsoft Excel and Matlab. In each trial, the first 3-5 consecutive cycles without stepping at each frequency were considered 'transition state' periods, indicating reactive postural responses. In the last half of the trial, a series of 3-5 consecutive cycles without stepping for the lowest frequency and 8–10 consecutive cycles without stepping for the remaining frequencies were considered the 'steady state' period where anticipatory postural adjustments were likely to occur (Figure 2) (Mills and Sveistrup, 2018; Bugnariu and Sveistrup, 2006).

EMG processing was performed in BioProc for Windows software (Robertson, 2008). Bias in the EMG signals was removed where appropriate and were fullwave rectified. EMG signals were not filtered; postural muscle onset latencies were determined from the raw signals. The first burst of activity associated with a perturbation, lasting more than 50ms and greater than two standard deviations above the within trial baseline (i.e., quiet period with no activity), indicated postural muscle onset latency activity (Mills and Sveistrup, 2018). Baseline was determined for each trial during the quiet stance period before platform movement initiation. To be included in the calculations of group muscle activity, responses had to be present in at least 30% of the directionally specific perturbations at each frequency (i.e., anterior muscles for backward perturbations, posterior muscles for forward perturbations) for transition state

[1] For the purposes of this thesis joint kinematics were not analysed, but may be investigated in subsequent analysis.

periods, and 50% for steady state periods. For the 0.1 Hz frequency, the recruitment threshold was reduced to 20% of perturbations for both transition and steady state (Mills and Sveistrup, 2018; Bugnariu and Sveistrup, 2006). Considering that each cycle had a different duration dependent on frequency, muscle onset latencies are presented as a percentage of half-cycle time, which is the movement of the platform from one extreme position to the other extreme position. If muscle activity began after zero, latencies were positive indicating reactive responses. If muscle activity began before zero, latencies were negative indicating anticipatory responses (Bugnariu and Sveistrup, 2006).

Tonic postural muscle activity was determined as follows: in each trial, a period of inactivity (i.e., no muscle bursts occurring), was identified in transition and steady state. For each participant, this was then compared to their 'baseline' which was determined during a period in steady state in NCT at 0.1 Hz where no burst activity was present. Tonic activity for each trial was then expressed as a percentage of the baseline tonic activity level in NCT steady state at 0.1 Hz.

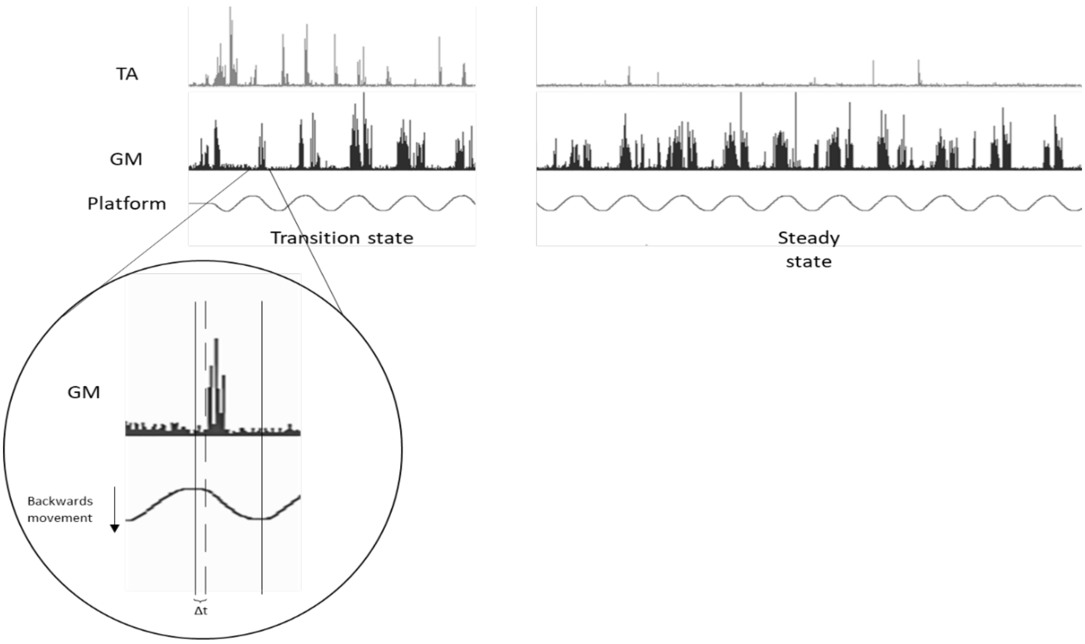

Figure 2. Perturbation protocol depicting platform oscillation at 0.25 Hz with corresponding EMG signals from tibialis anterior (TA) and gastrocnemius medialis (GM) during transition and steady states. The enlarged EMG signal of the GM is an example of one backwards platform movement (1/2 cycle), indicated by the two black vertical lines: the first black line indicates the start of the backwards movement and the second indicates the end of the backwards movement and the beginning of the forward movement. The dashed line indicates the start of the muscle activation and the time between the first black line and the dashed line (Δt) is the muscle onset latency.

2.5 Statistical Analysis

Descriptive analysis was used to summarize the participant demographics and stepping data. IBM SPSS Statistics version 26 was used to analyse tonic and bursting muscle activity. After considering skewness and kurtosis, conducting Shapiro-Wilk tests of normality and homogeneity of variance using a Levene's test, the onset latency data were determined to be parametric in all muscles, apart from the TA, while tonic activity was determined to be non-parametric for all muscles. To determine main effects and interactions between the different frequencies, conditions and states (i.e. to address hypotheses i, ii, and iii), a 3 (frequencies) x 3 (conditions) x 2 (transition and steady state period) factorial ANOVA was used for the parametric data, while Bonferroni post hoc tests were performed to identify significant differences in pairwise comparisons. Accepted level of significance was set at $p < 0.05$. For the tonic activity and the TA onset latencies which were determined as non-parametric, the same 3x3x2 ANOVA was used to determine significant differences, while Wilcoxon sign ranked tests were run for the pairwise comparisons where main effects or interactions were observed. To correct for multiple tests, significance level was adjusted such that α was determined by dividing 0.05 by the number of comparisons. Cognitive data were analysed using a Friedman test to compare the cognitive errors between frequencies and conditions, while Wilcoxon sign ranked tests were run for the pairwise comparisons that main effects or interactions were observed. To correct for multiple tests, significance level was adjusted such that α was determined by dividing 0.05 by the number of comparisons.

3. Results

3.1 Stepping responses

The total number of steps taken, the number of participants who stepped and the ranges of steps taken immediately after the platform started oscillating at each frequency and each condition are presented in Table 1. Stepping data were not statistically analysed and the numbers presented are raw counts. Regardless of condition, no steps were taken at 0.1 Hz. In CB, a single participant took 6 steps at 0.25 Hz, while 13 participants took a total of 68 steps at 0.5 Hz which was the highest number of steps recorded. In ST, 2 participants took 8 steps at 0.25 Hz and 7 participants took a combined 39 steps at 0.5 Hz. The least number of steps taken were recorded in the NCT at 0.25 Hz, with 2 participants taking 5 steps, while at 0.5 Hz a total of 25 steps were taken by 7 participants. Though not presented in Table 1, some falls were recorded: four participants stepped off the platform when they were introduced to the 0.5 Hz perturbation for the first time. Trials were then stopped and restarted to allow participants to safely step back on to the platform.

Table 2. Stepping responses at each frequency and condition.

	Frequency		
	0.1 Hz	0.25 Hz	0.5 Hz
CB	-	**6**/1 (6)	**68**/13 (1-10)
ST	-	**8**/2 (3-5)	**39**/7 (2-14)
NCT	-	**5**/2 (1-4)	**25**/7 (1-7)

Bold text indicates the total number of steps taken, followed by the number of participants who stepped immediately after the initiation of platform oscillation at each frequency. The numbers presented in parentheses indicate the ranges of steps taken.

3.2 Cognitive errors

The total number of cognitive errors, the number of participants who made errors and the range of errors made during the trial duration for each frequency and condition are presented in Table 2. No errors were made in the ST, regardless of frequency, while in CB, the fewest errors were made during quiet stance. A Friedman test revealed significant interaction effect between cognitive tasks and frequencies ($\chi^2(7) = 101.167$, $p < 0.001$). Post hoc analysis with

Wilcoxon signed-rank tests was conducted (adjusted α < 0.003) and showed significant differences between CB and ST at 0.1 Hz (Z= -3.633, p< 0.001, mean score= 0), at 0.25 Hz (Z= -3.635, p< 0.001, mean score= 0) and at 0.5 Hz (Z= -3.739, p< 0.001, mean score= 0). Significant differences were also found for the CB task between quiet stance and 0.1 Hz (Z= -3.184, p= 0.001, mean score= 9.57), 0.25 Hz (Z= -3.152, p= 0.002, mean score= 9.5) as well as 0.5 Hz (Z= -3.4, p= 0.001, mean score= 9.94). No significant differences were found for the ST between frequencies.

Table 3. Cognitive errors at each frequency and condition

		Frequency		
	Quiet Stance	0.1 Hz	0.25 Hz	0.5 Hz
CB	**19**/8 (1-3)	**74**/17 (1-12)	**72**/17 (1-13)	**73**/18 (1-9)
ST	-	-	-	-

Bold text indicates the total number of cognitive errors made, followed by the number of participants who made mistakes during the trial at each frequency and condition. The numbers presented in the parentheses indicate the ranges of errors made.

3.3 EMG

3.3.1 Onset latencies

For the Tibialis Anterior, significant frequency effects were found (F(2,95)= 17.134, p< 0.001, η_p^2= 0.265) and a Wilcoxon signed ranks test revealed significant difference (adjusted α < 0.017) between 0.25 Hz and 0.5 Hz (Z= -4.22, p< 0.001, mean score= 11.36) with onset latencies occurring earlier at 0.5 Hz (M= -2.7, SD= 8.1% half cycle). Significant interaction effects between frequency and state were also found (F(2,95)= 3.514, p= 0.034, η_p^2=0.069). A Wilcoxon signed rank test showed significant differences in transition state (adjusted α < 0.008) between 0.25 Hz and 0.5 Hz (Z= -3.645, p< 0.001, mean score= 10.93) with onset latencies occurring earlier at 0.5 Hz (M= 4.8, SD= 1% half cycle) compared to 0.25 Hz (M= 14, SD= 2.2% half cycle) (Figure 3A). No main effects were found for the Quadriceps.

Significant frequency effects were found for the Gastrocnemius Medialis ($F(2,223)= 68.410$, $p< 0.001$, $\eta_p^2= 0.38$) with post hoc comparisons identifying significant differences between 0.1 Hz and 0.25 Hz ($p< 0.001$), with onset latencies occurring earlier at 0.1 Hz (M= -4.46, SD= 1% half cycle) compared to 0.25 Hz (M= 9.18, SD= 2.5% half cycle). Significant differences were also observed between 0.25 Hz and 0.5 Hz ($p<0.001$) with muscle onset latencies occurring earlier at 0.5 Hz (M= -9.54, SD= 3.3% half cycle), and between 0.1 Hz and 0.5 Hz ($p= 0.003$) with muscle onset latencies occurring earlier at 0.5 Hz. A significant interaction effect between frequency and state was also found for the Gastrocnemius Medialis ($F(2,223)= 4.244$, $p= 0.016$, $\eta_p^2= 0.037$), and post hoc comparisons revealed that transition state was significantly different to steady state at 0.25 Hz ($p= 0.001$), with onset latencies occurring earlier in steady state (M= 5.17, SD= 1.6% half cycle) than in transition state (M= 13.19, SD= 2.2% half cycle) (Figure 3B).

Significant frequency effects were also seen in the Bicep Femoris ($F(2,100)= 8.944$, $p< 0.001$, $\eta_p^2= 0.152$). Post hoc comparisons identified significant differences between 0.1 Hz and 0.25 Hz ($p< 0.001$), with onset latencies occurring earlier at 0.1 Hz (M= -2.9, SD= 1.8% half cycle) compared to 0.25 Hz (M= 9.5, SD= 4.6% half cycle). Significant differences were also seen between 0.25 Hz and 0.5 Hz ($p< 0.001$), with onset latencies occurring earlier at 0.5 Hz (M= -1.34, SD= 3.9% half cycle) (Figure 3B).

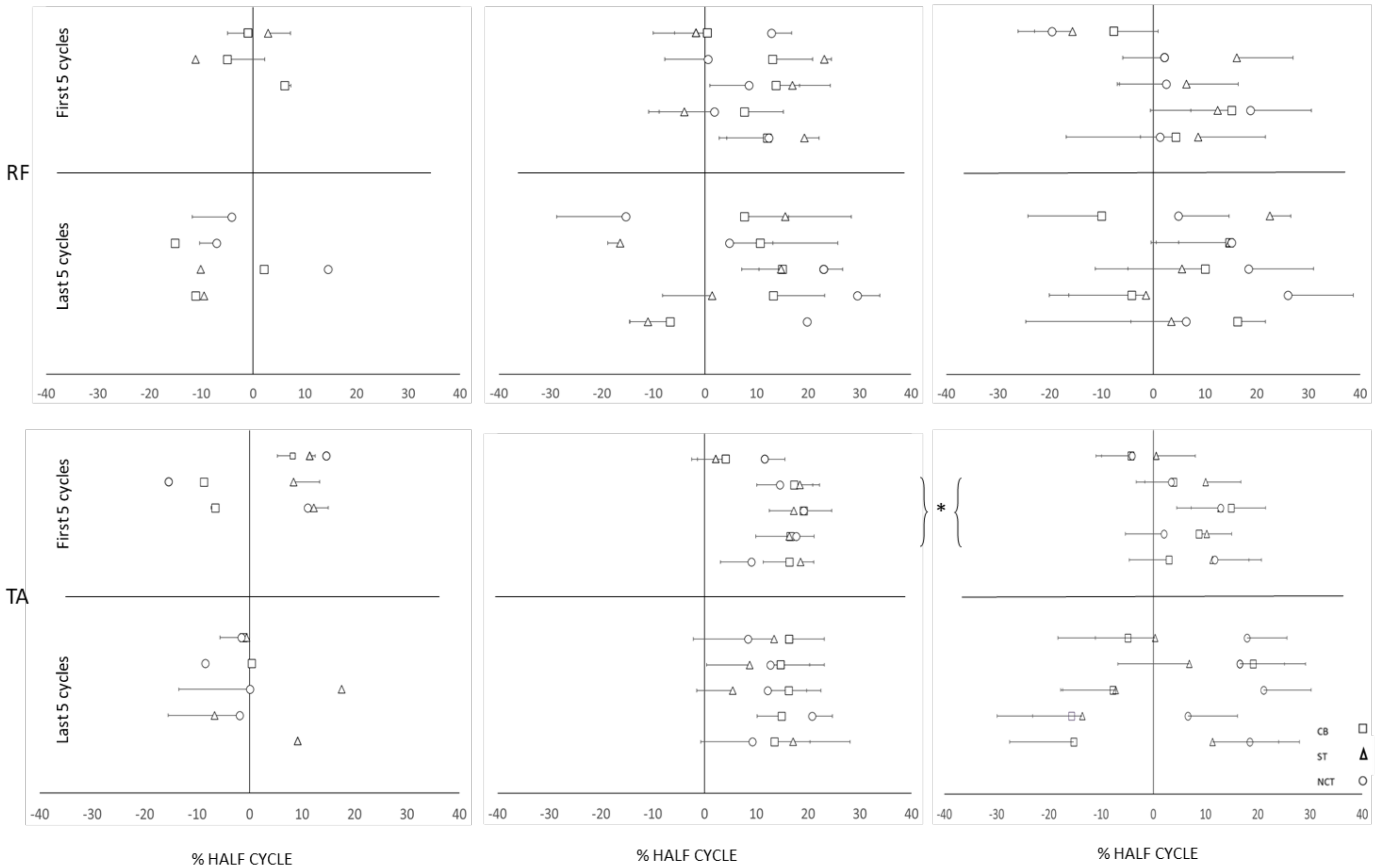

Figure 3A. Postural muscle onset latencies (mean ± SE) during the first 5 and last 5 cycles for the Rectus Femoris and Tibialis Anterior in each frequency and condition. Onset latencies are expressed as a percentage of half cycle time perturbations. Results from counting backwards (CB), Stroop test (ST) and no cognitive task (NCT) conditions are represented by squares, triangles, and circles, respectively. Zero (0) represents the time at which the platform changed direction; the platform begins to slow down at the 50% half cycle mark. Where latencies begin after zero (0), reactive responses are indicated by positive values. Where muscle activity begins before zero, latencies are negative, indicating anticipatory responses.

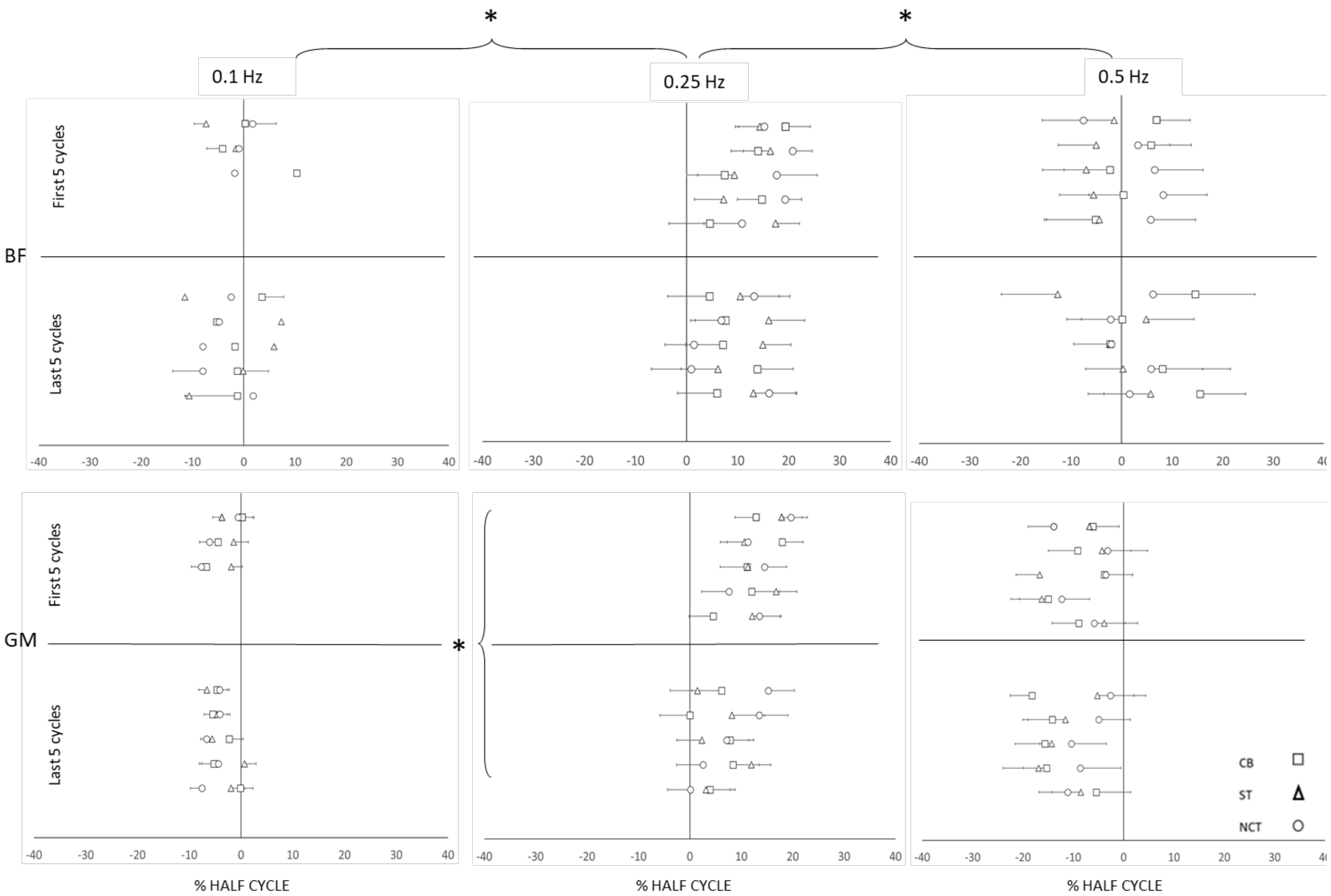

Figure 3B. Postural muscle onset latencies (mean ± SE) during the first 5 and last 5 cycles for the Bicep Femoris and Gastrocnemius Medialis in each frequency and condition. Onset latencies are expressed as a percentage of half cycle time perturbations. Results from counting backwards (CB), Stroop test (ST) and no cognitive task (NCT) conditions are represented by squares, triangles, and circles, respectively. Zero (0) represents the time at which the platform changed direction; the platform begins to slow down at the 50% half cycle mark. Where latencies begin after zero (0), reactive responses are indicated by positive values. Where muscle activity begins before zero, latencies are negative, indicating anticipatory responses.

No main effects were identified between dual and single tasking for any muscle. Figure 4 illustrates the onset latencies of each muscle, at all three frequencies and conditions, as well as transition and steady state.

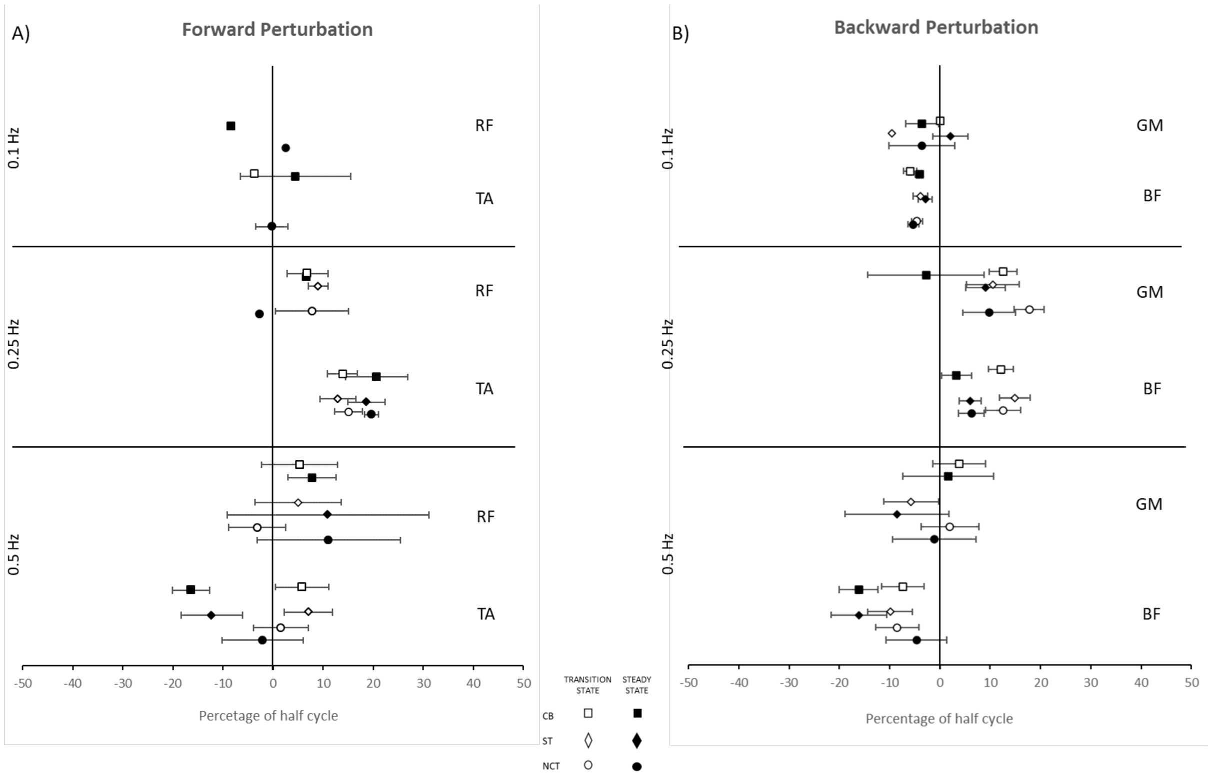

Figure 4. Postural muscles onset latencies (mean ± SE) during forward (A) and backward (B) perturbations at the three frequencies of platform oscillation. Onset latencies are expressed as a percentage of half cycle time for muscles normally associated with forward (TA and RF in panel A) or backward (GM and BF in panel B) perturbations. Results from transition and steady states are represented by open and filled icons, respectively, while counting backwards (CB), Stroop test (ST) and no cognitive task (NCT) conditions are represented by squares, diamonds, and circles, respectively. Zero (0) represents the time at which the platform changed direction; the platform begins to slow down at the 50% half cycle mark. Where latencies begin after zero (0), reactive responses are indicated by positive values. Where muscle activity begins before zero, latencies are negative, indicating anticipatory responses. Transition and steady state icons are offset for clear visual presentation.

3.3.2 Tonic activity

Tonic postural muscle activity was expressed as a percentage of the baseline tonic activity level in NCT steady state at 0.1 Hz. No main effects were found for tonic activity in the Tibialis

Anterior, while significant frequency effects were found for tonic activity in the Rectus Femoris ($F(2,339)= 4.474$, $p= 0.012$, , $\eta_p^2= 0.026$). Wilcoxon signed rank tests revealed significant differences (adjusted $\alpha < 0.017$) between 0.1 Hz and 0.5 Hz (Z= -3.862, $p< 0.001$, mean score= 62.11), with tonic activity being greater at 0.5 Hz (M= 177% baseline, SD= 79.2), as well as between 0.25 Hz and 0.5 Hz (Z= -3.468, $p= 0.001$, mean score= 61.21) where tonic activity was again greater at 0.5 Hz compared to 0.25 Hz (M= 144.1% baseline, SD= 29.5) (Figure 5).

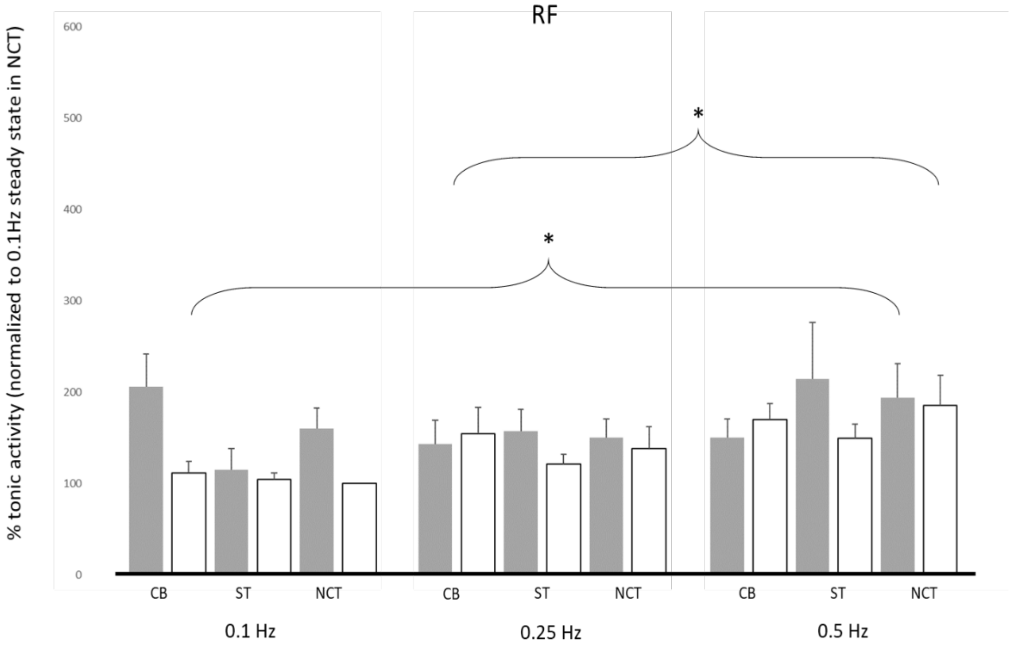

Figure 5. % Tonic activity for the Rectus Femoris (mean ± SE) for each frequency, condition, as well as transition and steady state, represented by grey and white colour respectively. The asterisks (*) indicate significant difference between frequencies

Significant frequency effects were also found for tonic activity in the Gastrocnemius Medialis ($F(2,338)= 4.168$, $p= 0.016$, , $\eta_p^2= 0.024$). Wilcoxon signed rank tests showed significant differences (adjusted $\alpha < 0.017$) between 0.25 Hz and 0.5 Hz (Z= -2.776, $p= 0.006$, mean score= 61.98), with tonic activity being greater at 0.5 Hz (M= 274.8% baseline, SD= 201.4) compared to 0.25 Hz (M= 184.4% baseline, SD= 77.5) (Figure 6).

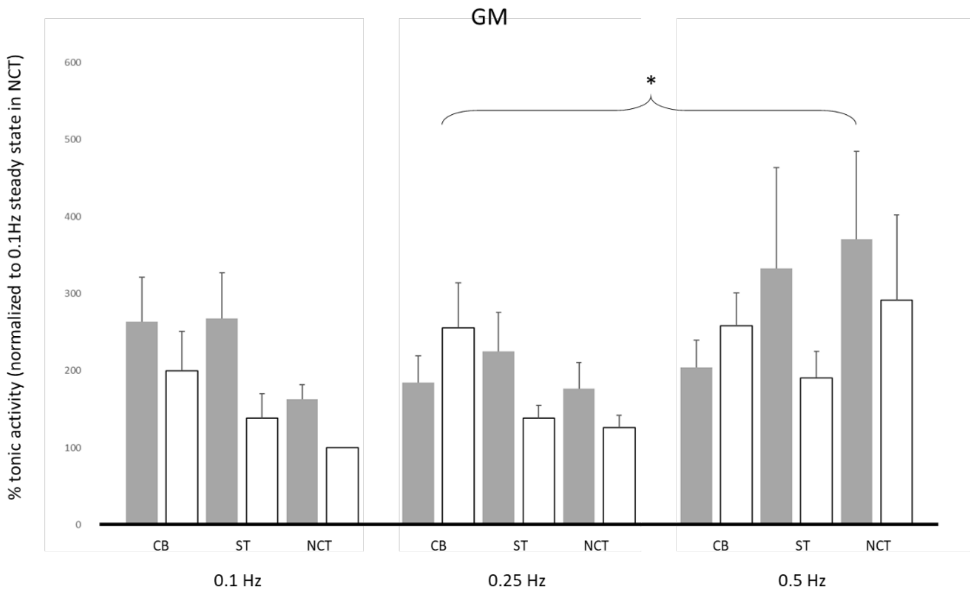

Figure 6. % Tonic activity for the Gastrocnemius Medialis (mean ± SE) for each frequency, condition, as well as transition and steady state, represented by grey and white colour respectively. The asterisk (*) indicates significant difference between frequencies.

For the tonic activity in the Bicep Femoris, significant interaction effects between frequency and condition were found (F(4,339)= 2.488, p= 0.043, , η_p^2= 0.029). Wilcoxon signed rank tests identified significant differences (adjusted α < 0.0166) in NCT between 0.25 Hz and 0.5 Hz (Z= -2.466, p= 0.014, mean score= 20.79) with tonic activity being greater at 0.5 Hz (M= 291% baseline, SD= 3.5) compared to 0.25 Hz (M= 127% baseline, SD= 21.9) (Figure 7). No main effects were identified between dual and single tasking for any muscle.

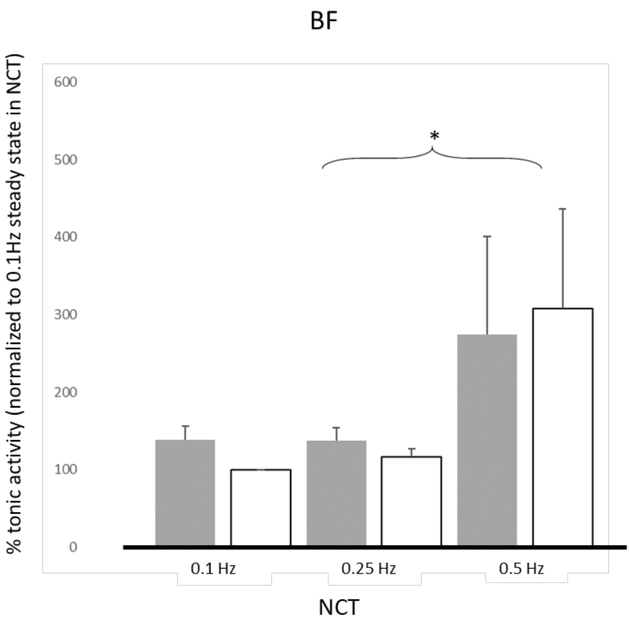

Figure 7. % Tonic activity for the Bicep Femoris (mean ± SE) during NCT for each frequency as well as transition and steady state, represented by grey and white colour respectively. The asterisk (*) indicates significant difference between frequencies.

4. Discussion

The aim of this study was to identify whether dual tasking would affect young adults' ability to generate anticipatory postural adjustments. An oscillating platform, moving at three frequencies (0.1 Hz, 0.25 Hz and 0.5 Hz), was used to assess postural responses as expressed by muscle onset latencies and tonic activity during dual tasking and single tasking. Two different cognitive tasks (counting backwards and Stroop task) were also used in order to identify if the different nature of the tasks would affect postural responses.

The first hypothesis was that participants would take more steps, make more cognitive errors, and display delayed postural muscle onset latencies, as well as greater tonic activity during dual tasking compared to the single tasking condition. This hypothesis was partially supported, as fewer steps were taken in the NCT (Table 1), since most (if not all) available attentional resources should be allocated for the recovery and maintenance of balance. As the frequencies increased, the number of steps taken also increased, in both dual tasking and single tasking conditions. The increased number of steps taken in the higher frequencies reflects the increasing difficulty, and thus the increased attentional demands of the postural task. This was also observed in a single-tasking study, where adolescents were exposed to repeated anterior-posterior platform perturbations (Mills and Sveistrup, 2018). The authors found that a greater number of steps was taken at the higher frequencies with 26 steps taken at 0.5 Hz and 11 steps taken at 0.61 Hz. The greater number of steps recorded in the present study is likely due to a slight difference in perturbation protocol between the two studies. While Mills and Sveistrup (2018) made use of sinusoidal oscillations with incrementally increasing frequencies within a single trial, the current study protocol used separate perturbation trials at each frequency due to equipment limitations. Therefore, the participants in this study experienced larger perturbations since their starting point was quiet stance (i.e., 0 Hz), compared to the participants in the former study who experienced increases of 0.15 Hz, 0.25 Hz, and 0.11 Hz as the platform moved from 0.1 Hz to 0.25 Hz, 0.25 Hz to 0.5 Hz, and 0.5 Hz to 0.61 Hz, respectively. The larger perturbations, along with the additional cognitive demands of the dual tasking, could explain why the participants in this study took more than double the number of steps compared to Mills and Sveistrup (2018).

As expected, more cognitive errors were also observed in dual tasking when the CB task was performed, compared to the baseline errors made in quiet stance (single tasking) (Table 2).

However, the number of errors made between frequencies during dual tasking was not significantly different, possibly indicating that increased postural task difficulty did not have an additional impact on cognitive performance of the CB task. This lack of discrimination between the different frequencies could be attributed to the fact that the reaction time of the CB performance was not measured in this study. It could be the case that at 0.1 Hz, participants performed more, quicker calculations but still had a high number of errors, while at 0.5 Hz, they performed slower, and fewer subtractions but with a similar number of errors. Since reaction time was not measured, we are unable to conclude that the increased difficulty of the postural task did not indeed have an impact on cognitive performance. In regard to ST performance, no errors were made during dual tasking, nor in single tasking. This could be a reflection of the difficulty level of the ST. Each word appeared on the screen every 4s; if words were to change more frequently instead (e.g., every 1s), hence increasing difficulty, differences might have been observed between dual tasking and single tasking in ST performance.

4.1 High frequency perturbations are compensated through increased tonic activity and earlier muscle activation.

As part of the first hypothesis, tonic activity and onset latencies were expected to be different between dual tasking and single tasking. However, the results did not indicate that. No significant differences were identified between dual and single tasking (in CB nor ST) at each of the three frequencies. However, significant differences in tonic activity and onset latencies were identified between frequencies when conditions were not considered. Higher tonic activity at 0.5 Hz was observed for the Gastrocnemius and Quadriceps, indicating that in the most destabilizing frequency, participants were able to control posture by adopting a functional joint stiffening method (Needle et al., 2014). This has previously been hypothesized in dual tasking studies under postural tasks of different difficulty levels (Albertsen et al., 2017; Dault et al., 2001). For instance, Dault et al. (2001) found that when participants performed the Stroop task and adopted a seesaw stance compared to shoulder width stance, mean sway frequency was increased, leading them to assume that stiffness was increased to deal with the increased cognitive demands. The same assumption was made by Albertsen et al. (2017) after reduced postural sway was found when participants were asked to keep their feet together in quiet standing, hence increasing difficulty, and count backwards by 7. However, EMG measures were not investigated in these studies to confirm their assumptions. In the current study, even though tonic activity was similar between frequencies in the two dual tasking conditions, for the Hamstrings in the NCT, tonic activity was significantly higher at 0.5 Hz (Figure 6) compared to the 0.25 Hz frequency. This prolonged muscle activation could be associated with an element of threat observed at this specific frequency. Studies have found that when the surface height is increased, therefore increasing postural threat, a stiffening strategy is adopted and more postural adjustments are made (Carpenter et al., 2001; Adkin et al., 2000). Therefore, the 0.5 Hz frequency could be perceived as a 'threatening' perturbation, leading to increased tonic activity. However, this was only evidenced in the single tasking condition. It has been reported that postural control is automated when attention is focused on the cognitive tasks, sometimes resulting in improved postural performance (Huxhold et al., 2006; Wulf et al., 2004; McNevin and Wulf, 2002). Therefore, during single tasking, attention would be focused on the maintenance of balance, causing "interference on automatically self-organized postural behaviour" (Bernard-Demanze et al., 2009; Wulf et al., 2001). This has been shown during quiet stance where explicit instructions were given to young adults towards focusing their attention to the postural task, resulting in increased body sway (Vuillerme and Nafati, 2007). Therefore, higher tonic activity of the Hamstrings observed in the NCT at 0.5 Hz could indicate that in the single tasking condition, participants had to direct their attention in the postural task, focusing to the 'threatening' factor of the high frequency perturbation, leading to a "stiffer" position. Considering that this was only evidenced for the Hamstrings, one could assume that they adopted a 'leaning forward' postural response causing the muscle to activate more than others in order to pull the body backwards. However, kinematic and kinetic measures were not investigated in the current study, therefore these assumptions cannot be confirmed.

A frequency effect was also observed for onset latencies regardless of condition. When participants were faced with the lowest frequency, they were able to activate their muscles earlier than at the 0.25 Hz frequency (Figure 3), possibly indicating that the 0.1 Hz frequency was an easy postural task. During platform oscillations at low frequencies, such as 0.1 Hz, postural sway, as expressed by mean velocity, is only slightly increased compared to quiet stance (Sakanaka et al., 2021). In fact, when dealing with low frequency sway (less than 0.1 Hz), stability can be maintained through the use of visual cues (Dichgans et al., 1976;

Lestienne et al., 1977), while when faced with frequencies greater than 0.1 Hz, ankle and feet proprioception is used to stabilise standing posture (Diener et al., 1984). This would suggest that in the current study, the lower frequency possibly only required minor corrections in COM sway, and this was dealt with by instant muscle activity. Interestingly, muscle onsets at 0.5 Hz occurred earlier than in the lower frequencies. Similar results were found by Azzi et al. (2017) in single tasking anticipatory balance, whereby large perturbations, caused by suddenly releasing a load attached to the participant's trunk, were compensated for by increasing magnitude and decreasing onset latency of muscle activation compared to lower perturbations. The onset of muscle activation seems to be controlled through anticipatory mechanisms brought about by a state of readiness or 'central set' (Jacobs and Horak, 2007). A central set has been defined by Prochazka (1989) as "a state of readiness to receive a stimulus or make a movement, represented by a task-dependent preparatory neural discharge within the central nervous system". It might be assumed then, that in the current study, sensory afference signalling a large perturbation triggered pre-determined postural responses to attend the anticipated requirements of the specific perturbation speed. Bugnariu and Sveistrup (2006) observed muscle activity to occur at around -50% half cycle for their young adults during single tasking, while in this study the overall timing of muscle activity for dual tasking and single tasking was around -5% half cycle for the Hamstrings and Gastrocnemius (forward perturbation) and around +10% half cycle for the Quadriceps and Tibialis Anterior (backwards perturbation) (Figure 4). This is possibly due to the perturbation protocol used in this study. Bugnariu and Sveistrup (2006), just like Mills and Sveistrup (2018) discussed above, incrementally increased the frequency of the perturbation up to 0.61 Hz within the same trial, while in this study the platform was perturbed at one of the three frequencies each time. Therefore, since the participants in this study were faced with larger perturbations, the increased cognitive demands of the postural task could delay their muscle activation compared to the young adults of Bugnariu and Sveistrup (2006).

4.2 The influence of perturbation intensity on anticipatory mechanisms.

The second hypothesis that participants would not be able to shift from reactive to anticipatory mechanisms during dual tasking was not supported by the results. The different conditions did not seem to have an effect, and when all three are combined, transition and steady state was only significantly different for the Gastrocnemius muscle at 0.25 Hz (Figure

3B). It has previously been found that compared to older adults, young adults are able to activate their muscles earlier within the first 3-5 cycles (transition state) of a new frequency (Bugnariu and Sveistrup, 2006), possibly due to prior experience gained throughout trials (Kennedy et al., 2013; van Ooteghem et al., 2009; Nashner, 1976). This was observed in the current study for the Gastrocnemius muscle at 0.25 Hz only. In the first 5 cycles, participants relied on reactive mechanisms and muscle activation was delayed. As they became accustomed to the perturbation speed, a shift to anticipatory mechanisms was observed, evidenced by earlier muscle activations in steady state. Muscle activations at 0.1 Hz did not follow any defined trend (e.g., earlier, or delayed activations), rather there appears to be no specific temporal organisation. As mentioned above, the 0.1 Hz frequency is likely not sufficiently threatening to balance and hence there is no immediate requirement to adapt from transition to steady state. When the speed is threatening to balance at 0.5 Hz, anticipatory mechanisms were triggered at transition state in the Hamstrings and Gastrocnemius muscles (Figure 3B), and then carried to steady state in order to maintain balance. It has previously been shown that when participants were faced with high frequency platform anterior-posterior translations (0.6 Hz) during single tasking with eyes open, no adaptations occur as ‘steady state muscle activity’ (i.e., earlier muscle activation) occurs within the first or second perturbation cycle (Sozzi et al., 2016). This seems to be the case in this study, since as observed in Figure 3B, early muscle activations in the Hamstrings and Gastrocnemius occurred in the first cycle. This indicates that the highest frequency required an urgent response from the postural control system to maintain balance.

4.3 Postural control is automated in dual and single tasking.

The last hypothesis was that onset latencies and tonic activity would be different between dual-tasking conditions (CB and ST). This hypothesis was made on the basis that the CB task does not require the use of external visual cues for its execution but rather turns one’s focus internally, negatively impacting postural control, while the ST is a visuo-verbal task that focuses one’s attention externally, improving balance (Jamet et al., 2004). However, such differences were not identified. Tonic activity and onset latencies were similar between the two cognitive tasks, as well as single tasking, possibly indicating that participants’ postural control was functioning on an automatic level in all conditions, including single tasking, and the errors observed in the CB task reflect the difficulty of the task itself. However, since no errors were observed in the ST, therefore assuming it is an ‘easy’ task in relation to the CB task as discussed above, we cannot identify whether the difference in cognitive performance of the two tasks was due to the difficulty level, or the differing nature of the tasks (i.e., external-internal focus).

5. Limitations

One of the limitations of this study is related to the power achieved. In order to achieve 80% power, a sample size of 35 participants would be required. However, because of interruptions due to the Covid-19 pandemic, only 20 participants were recruited; a post hoc analysis revealed that the power achieved was 54%. As seen in the results section, the effect of frequency on tonic activity for the Gastrocnemius Medialis, Bicep Femoris and Rectus Femoris ($\eta_p^2 = 0.024$, $\eta_p^2 = 0.029$, $\eta_p^2 = 0.026$, respectively) was small, considering an η_p^2 of 0.01 is a small effect, an η_p^2 of 0.06 a medium and an η_p^2 of 0.14 a large effect (Richardson, 2011). On the other hand, the effect of frequency on onset latencies for the Tibialis Anterior, Gastrocnemius Medialis and Bicep Femoris was large ($\eta_p^2 = 0.265$, $\eta_p^2 = 0.38$, $\eta_p^2 = 0.152$, respectively), showing that even with a small sample size, frequency had a significant effect on muscle onset latencies. Increasing the sample size might allow for differences between transition and steady sates to be identified and/or a larger effect of frequency on tonic activity to be found.

Increasing sample size could also allow for differences between dual tasking conditions to be identified, though this may be better explained by another limitation of the study, the dual-tasking protocol. Reaction time of the cognitive task performance during single tasking and dual tasking was not calculated. Cognitive performance was only based on the correct answers of the CB and ST, however measuring the reaction time of their answers would be a good indicator of attention allocation during the dual tasking conditions. It was observed that as the frequency of the platform perturbation increased, the number of cognitive errors for the CB was similar, however, had the reaction time of their answers been measured, a speed-accuracy trade-off could have been seen. It is unknown whether participants focused their attention on doing the task right or doing it fast (since the instructions given were 'count as accurately and as fast as you can'), and it has been found that they can adapt their speed and accuracy between trials and conditions (Reuss et al., 2015), therefore future studies should take this into account.

Kinematics and kinetics were not collected in this study. Collecting joint kinematic data and investigating COP and COM parameters, such as COP sway, range, velocity and COM displacement would have offered valuable information regarding the balance strategies used by the participants. When the perturbation can be dealt with using a feet-in-place strategy, either an ankle or hip strategy is used to counteract the perturbation (Horak et al., 1986). As the difficulty of the task increases, for instance when the perturbation frequency increases, a mixed ankle and hip strategy is seen, evidenced by combined ankle plantarflexion, knee and hip flexion (Runge et al., 1999). Measuring horizontal ground reaction forces (Gruben and Boehm, 2012), changes in angular momentum (Halvorsen, 2010) and trunk lean angle (Versteeg et al., 2016) are some ways of identifying whether a hip strategy is used. Another way of detecting what postural strategy is adopted is to investigate the cross-correlation of the trajectory of different segments (i.e., head-ankle, head-hip, hip-ankle) (Mills and Sveistrup, 2018). In adolescents, the use of the ankle strategy when the perturbation is not challenging is demonstrated by coupling of body segments, while when the perturbation frequency is threatening to balance, the use of a hip strategy is demonstrated by correlation of the ankle and hip and temporally displaced head and ankle segments (Mills and Sveistrup, 2018). In this study, it is assumed that a hip strategy was used during in the NCT condition since increased tonic activity was found for the Hamstrings at the highest frequency, possibly indicating forward lean of the trunk. Since no information regarding participants' joint kinematics and kinetics were obtained, it is unknown whether that is indeed the case or whether a combination of ankle and hip strategy was adopted.

Kinetic information, such as COP displacement and its location within the base of support, would also provide valuable insight into participants' postural control and their ability to avoid stepping or falling during perturbed balance dual tasking. Bugnariu and Sveistrup (2006) observed that in response to anterior-posterior perturbations during single tasking, older adults' COP was located at the extremes of their base of supportfor a longer period of time, compared to young adults, and particularly in transition state at high frequency perturbations, indicating less stability during the first 5 perturbation cycles. It would be useful therefore, to explore the effects of dual tasking on COP parameters, especially during transition and steady state, to identify whether young adults keep their COP within the safer regions of the boundaries of their33or not, and whether that happens within the first few cycles of the perturbation.

6. Future research and conclusion

Future studies should aim to address the limitations above and quantify the discrepancy of postural responses during dual tasking by investigating cross-correlation and time lag between body segments. Recently, Sozzi et al. (2020) found that the execution of a visual task (i.e., reading a text fixed to the moving platform or fixed to the ground) during high frequency anterior-posterior perturbations, increased the displacement of the head, but not the hip in young adults, indicating that priority was given to the stabilization of the pelvis over stabilization of the head. The authors also observed this to be at the expense of the visual task. It would be important therefore, to investigate whether a cognitive task that does not require the use of vision would elicit similar results, providing an insight into the interaction of sensory inputs and cognitive tasks on postural responses.

In conclusion, our results indicated that dual tasking does not influence the generation of anticipatory postural adjustments, however perturbation intensity does. In the high frequency perturbation, anticipatory mechanisms are generated sooner compared to the lower frequencies, as evidenced by earlier muscle onsets latencies, and a stiffer position seems to be adopted, as expressed by increased tonic activity at 0.5 Hz. Regarding the transition and steady state of the perturbations, the lower frequency (0.1 Hz) seemed too easy to require adaptation and the highest frequency (0.5 Hz) was proved large enough to trigger earlier muscle activation from transition state which was then carried to steady state. Since the postural characteristics measured remained unchanged during single and dual tasking, it is assumed that postural control was automated, and the cognitive errors observed in the two tasks reflect their difficulty level. Future studies should aim to investigate the balance strategies adopted by exploring body kinematics, as well as take into account the reaction time of the cognitive task to better understand participants’ allocation of attention during perturbed balance dual tasking.

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Appendix 1 The percentage of activity bursts per condition by each participant in transition and steady state for each muscle. The – indicates no bursts while the x indicates that no data were available for analysis.

Tibialis Anterior

		0.1CB	0.1ST	0.1NCT	0.25CB	0.25ST	0.25NCT	0.5CB	0.5ST	0.5NCT
P1	TS	-	-	20%	80%	10%	20%	10%	20%	-
	SS	5%	5%	10%	25%	5%	5%	5%	20%	30%
P2	TS	-	-	-	-	10%	50%	60%	40%	70%
	SS	-	5%	10%	5%	5%	15%	25%	-	30%
P3	TS	-	-	-	20%	10%	-	80%	40%	40%
	SS	-	-	-	-	-	5%	-	-	-
P4	TS	10%	10%	-	80%	50%	70%	50%	50%	40%
	SS	-	5%	15%	30%	45%	25%	15%	15%	30%
P5	TS	-	-	-	60%	40%	20%	80%	100%	60%
	SS	-	-	-	30%	20%	-	15%	15%	25%
P6	TS	-	20%	-	50%	50%	50%	30%	50%	70%
	SS	-	5%	5%	30%	30%	15%	45%	30%	25%
P7	TS	-	-	-	60%	50%	20%	50%	20%	20%
	SS	10%	-	-	15%	10%	15%	10%	-	5%
P8	TS	10%	-	-	50%	-	-	50%	30%	60%
	SS	-	-	-	5%	5%	-	30%	30%	55%
P9	TS	-	-	-	50%	-	40%	50%	40%	90%
	SS	-	-	-	50%	-	50%	40%	45%	85%
P10	TS	10%	20%	-	40%	50%	20%	30%	50%	30%
	SS	-	-	-	10%	30%	25%	-	-	5%
P11	TS	-	-	10%	30%	30%	30%	20%	60%	90%
	SS	-	-	-	-	-	10%	10%	30%	25%
P12	TS	-	-	-	30%	30%	20%	90%	60%	50%
	SS	-	-	-	35%	30%	10%	10%	5%	5%
P13	TS	-	-	10%	30%	60%	40%	80%	100%	50%
	SS	-	-	-	5%	-	-	5%	20%	10%
P14	TS	30%	10%	10%	90%	50%	100%	100%	90%	70%
	SS	10%	-	10%	55%	75%	80%	70%	100%	75%
P15	TS	-	10%	-	20%	20%	10%	60%	50%	50%
	SS	5%	-	-	5%	20%	30%	30%	45%	45%
P16	TS	10%	-	-	50%	80%	50%	50%	50%	90%
	SS	-	-	-	20%	50%	-	50%	50%	65%
P17	TS	-	-	-	40%	30%	50%	80%	50%	40%
	SS	5%	-	-	25%	30%	15%	30%	70%	20%
P18	TS	-	-	-	40%	40%	20%	60%	90%	60%
	SS	-	-	-	15%	-	0%	45%	10%	40%
P19	TS	-	-	-	10%	10%	10%	80%	50%	10%
	SS	-	-	-	-	-	-	50%	-	-
P20	TS	-	-	-	10%	10%	20%	60%	40%	20%
	SS	-	5%	5%	-	5%	-	-	-	5%
MEAN	TS	4%	4%	3%	42%	32%	32%	59%	54%	51%
	SS	2%	1%	3%	18%	18%	15%	24%	24%	29%
SD	TS	7%	7%	6%	24%	22%	25%	24%	24%	26%
	SS	3%	2%	5%	17%	21%	20%	20%	27%	25%

Bicep Femoris

		0.1CB	0.1ST	0.1NCT	0.25CB	0.25ST	0.25NCT	0.5CB	0.5ST	0.5NCT
P1	TS	20%	10%	-	80%	20%	10%	10%	-	-
	SS	5%	-	5%	45%	15%	10%	20%	15%	15%
P2	TS	-	-	20%	60%	-	40%	-	40%	50%
	SS	-	15%	-	25%	20%	25%	-	70%	70%
P3	TS	20%	10%	10%	40%	50%	30%	80%	100%	90%
	SS	5%	-	5%	5%	10%	25%	80%	75%	100%
P4	TS	-	-	-	-	-	-	-	10%	-
	SS	-	-	-	-	-	-	-	-	-
P5	TS	-	-	10%	50%	40%	40%	60%	100%	80%
	SS	-	-	-	35%	30%	30%	90%	45%	85%
P6	TS	10%	20%	10%	20%	30%		10%	20%	40%
	SS	5%	10%	10%	-	15%	10%	5%	10%	15%
P7	TS	10%	20%	10%	70%	70%	60%	60%	60%	60%
	SS	-	-	-	65%	55%	50%	75%	45%	40%
P8	TS	-	10%	10%	80%	40%	10%	80%	100%	100%
	SS	-	-	5%	30%	40%	10%	35%	70%	75%
P9	TS	-	10%	-	10%	-	20%	30%	30%	20%
	SS	-	5%	5%	20%	-	20%	25%	20%	25%
P10	TS	-	-	-	-	-	-	-	-	10%
	SS	-	-	-	-	5%	-	5%	-	-
P11	TS	-	-	10%	20%	10%	30%	10%	20%	-
	SS	-	-	-	-	-	35%	5%	-	-
P12	TS	X	X	X	x	x	x	x	x	x
	SS	X	X	X	x	x	x	x	x	x
P13	TS	X	X	X	x	x	x	x	x	x
	SS	X	X	X	x	x	x	x	x	x
P14	TS	30%	20%	20%	50%	80%	70%	100%	100%	70%
	SS	30%	15%	10%	50%	70%	55%	75%	95%	80%
P15	TS	20%	-	10%	30%	40%	40%	70%	50%	60%
	SS	20%	5%	5%	50%	15%	25%	70%	40%	40%
P16	TS	-	-	-	50%	60%	40%	60%	90%	100%
	SS	-	-	-	40%	10%	20%	75%	80%	95%
P17	TS	10%	10%	-	20%	50%	40%	70%	100%	50%
	SS	-	-	-	10%	25%	50%	50%	65%	55%
P18	TS	-	10%	10%	80%	50%	50%	40%	40%	80%
	SS	-	10%	-	25%	25%	50%	25%	35%	50%
P19	TS	20%	30%	-	80%	90%	80%	60%	30%	30%
	SS	10%	5%	-	15%	45%	60%	65%	40%	40%
P20	TS	-	-	-	40%	30%	10%	70%	70%	50%
	SS	-	-	-	40%	25%	40%	25%	55%	40%
MEAN	TS	7%	8%	6%	39%	33%	29%	41%	48%	45%
	SS	4%	3%	2%	23%	20%	26%	36%	38%	41%
SD	TS	10%	9%	7%	30%	29%	25%	34%	39%	35%
	SS	8%	5%	3%	21%	20%	20%	33%	31%	34%

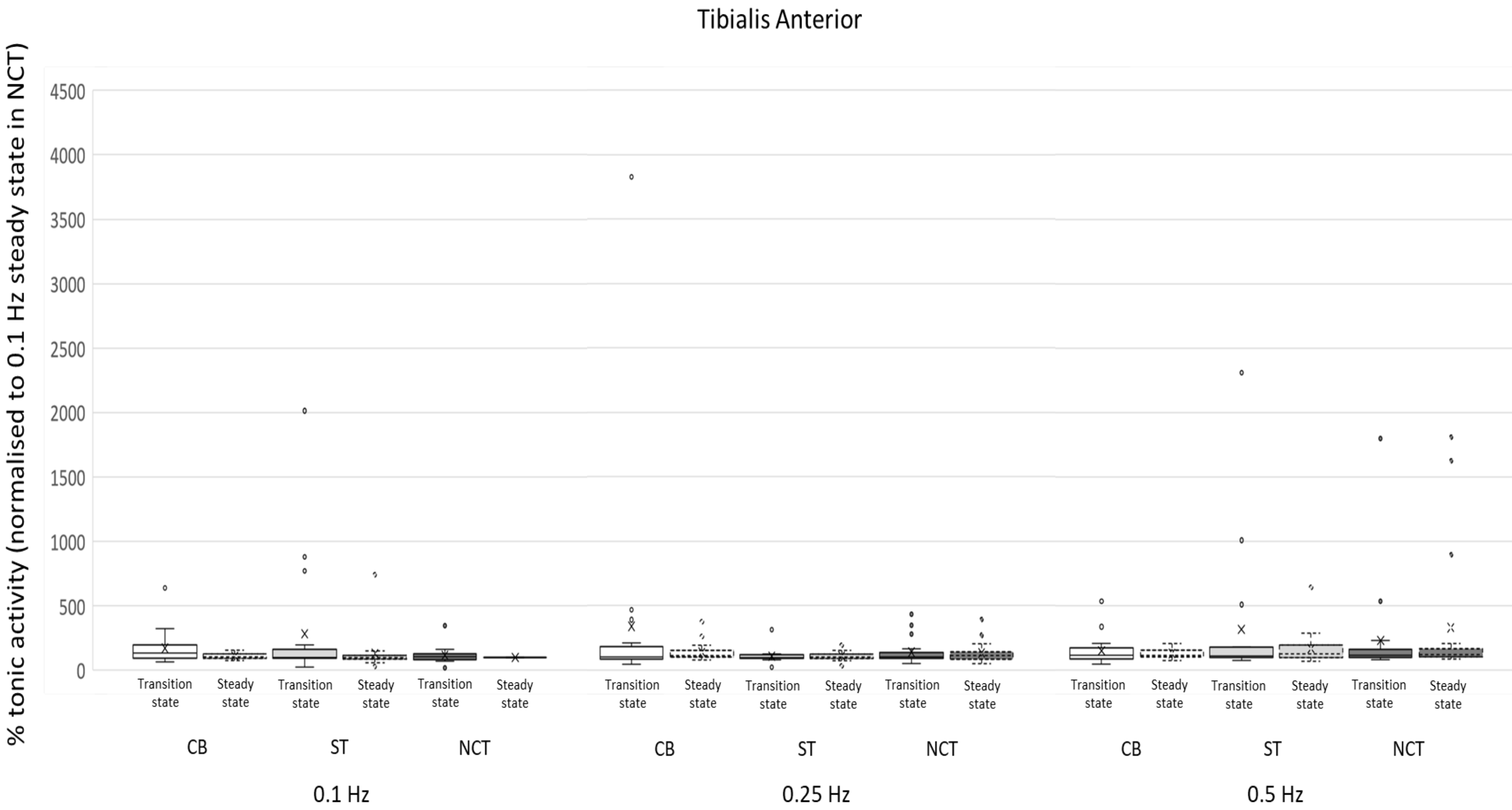
Gastrocnemius Medialis

		0.1CB	0.1ST	0.1NCT	0.25CB	0.25ST	0.25NCT	0.5CB	0.5ST	0.5NCT
P1	TS	-	-	-	10%	20%	-	30%	40%	10%
	SS	5%	-	-	5%	-	10%	35%	5%	40%
P2	TS	40%	10%	20%	70%	90%	80%	90%	80%	50%
	SS	10%	-	-	70%	45%	60%	65%	80%	45%
P3	TS	10%	40%	40%	100%	100%	70%	100%	100%	100%
	SS	20%	35%	50%	70%	95%	95%	100%	100%	100%
P4	TS	10%	10%	10%	20%	70%	80%	90%	80%	50%
	SS	10%	5%	15%	15%	45%	65%	65%	80%	45%
P5	TS	10%	30%	10%	100%	100%	100%	100%	100%	100%
	SS	20%	40%	25%	90%	100%	100%	100%	90%	95%
P6	TS	40%	50%	60%	80%	80%	90%	100%	90%	100%
	SS	40%	40%	35%	100%	80%	80%	100%	80%	90%
P7	TS	20%	40%	20%	80%	100%	100%	100%	100%	100%
	SS	15%	20%	10%	90%	85%	95%	95%	95%	90%
P8	TS	40%	30%	10%	100%	10%	10%	80%	100%	100%
	SS	25%	15%	10%	90%	40%	40%	30%	70%	70%
P9	TS	-	60%	-	-	-	30%	50%	40%	90%
	SS	-	45%	5%	-	-	45%	45%	45%	100%
P10	TS	50%	30%	10%	100%	80%	100%	90%	100%	100%
	SS	15%	25%	35%	95%	70%	80%	90%	90%	95%
P11	TS	10%	20%	20%	20%	80%	80%	70%	80%	90%
	SS	20%	10%	10%	10%	55%	60%	85%	75%	85%
P12	TS	20%	20%	-	90%	90%	90%	100%	70%	100%
	SS	-	5%	-	95%	90%	70%	90%	95%	95%
P13	TS	X	X	X	x	X	x	x	x	x
	SS	X	X	X	x	X	x	x	x	x
P14	TS	20%	10%	20%	90%	90%	90%	90%	100%	90%
	SS	30%	-	5%	65%	75%	65%	75%	70%	70%
P15	TS	20%	10%	20%	40%	40%	50%	80%	100%	90%
	SS	5%	10%	10%	10%	35%	35%	20%	75%	70%
P16	TS	-	50%	20%	90%	80%	70%	90%	100%	100%
	SS	-	15%	5%	75%	70%	40%	85%	65%	95%
P17	TS	10%	-	-	30%	70%	60%	100%	90%	50%
	SS	-	-	-	20%	35%	45%	80%	75%	30%
P18	TS	30%	30%	40%	90%	80%	70%	100%	100%	100%
	SS	40%	30%	15%	55%	50%	50%	95%	95%	100%
P19	TS	10%	50%	-	60%	90%	90%	80%	90%	90%
	SS	20%	10%	5%	30%	30%	80%	70%	80%	80%
P20	TS	30%	10%	40%	50%	50%	80%	60%	60%	40%
	SS	20%	-	15%	20%	35%	50%	75%	50%	55%
MEAN	TS	19%	25%	17%	61%	66%	67%	80%	81%	78%
	SS	15%	15%	13%	50%	52%	58%	70%	71%	73%
SD	TS	15%	19%	17%	36%	34%	32%	27%	27%	32%
	SS	13%	15%	14%	37%	31%	27%	29%	27%	28%

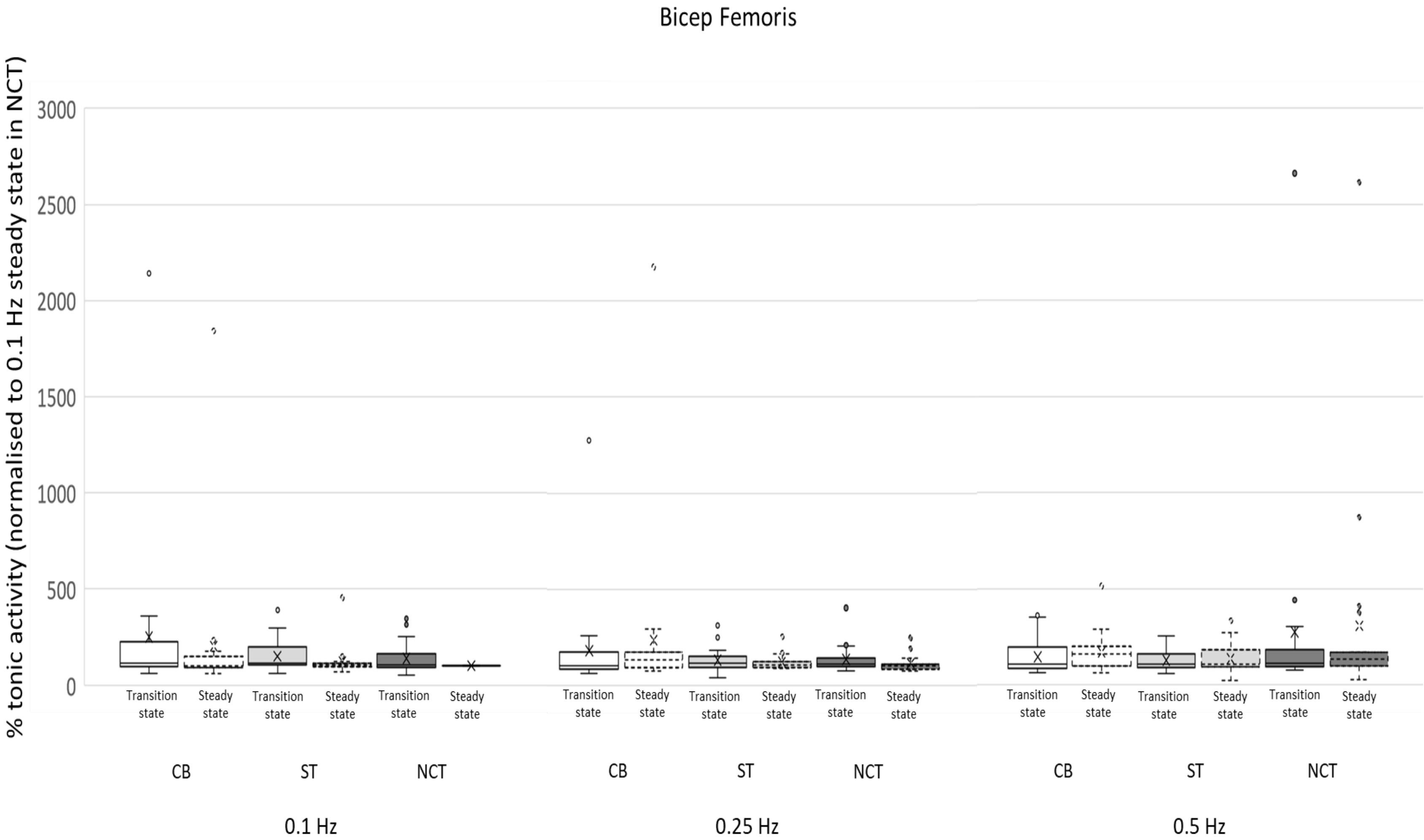
Rectus Femoris

		0.1CB	0.1ST	0.1NCT	0.25CB	0.25ST	0.25NCT	0.5CB	0.5ST	0.5NCT
P1	TS	-	-	-	20%	10%	20%	20%	-	-
	SS	-	-	-	-	-	-	5%	15%	10%
P2	TS	10%	10%	-	10%	-	40%	-	20%	60%
	SS	-	5%	-	-	-	25%	-	-	30%
P3	TS	20%	-	-	10%	-	10%	30%	40%	30%
	SS	-	-	-	-	-	15%	-	-	10%
P4	TS	-	10%	-	-	20%	-	10%	20%	20%
	SS	-	-	5%	10%	15%	-	-	-	5%
P5	TS	-	-	-	10%	-	10%	30%	40%	30%
	SS	-	-	-	-	-	15%	-	15%	20%
P6	TS	-	-	-	-	-	-	30%	20%	20%
	SS	-	-	-	-	-	-	15%	-	25%
P7	TS	-	-	-	-	10%	-	10%	-	10%
	SS	-	-	-	-	-	10%	-	-	-
P8	TS	-	10%	-	60%	-	40%	80%	60%	70%
	SS	-	5%	-	10%	10%	5%	55%	75%	75%
P9	TS	-	-	-	30%	-	30%	50%	20%	50%
	SS	-	-	-	15%	-	-	50%	25%	35%
P10	TS	10%	-	-	60%	80%	70%	50%	60%	40%
	SS	-	-	5%	75%	40%	45%	45%	55%	30%
P11	TS	-	-	-	20%	10%	20%	-	-	-
	SS	-	-	-	-	-	5%	-	5%	5%
P12	TS	-	-	-	60%	-	-	-	-	-
	SS	-	-	-	-	-	-	-	-	-
P13	TS	10%	10%	-	40%	-	10%	80%	40%	90%
	SS	-	-	-	15%	-	-	15%	15%	30%
P14	TS	10%	10%	-	50%	50%	60%	30%	100%	80%
	SS	25%	5%	15%	5%	35%	55%	10%	55%	60%
P15	TS	20%	-	-	40%	-	10%	40%	40%	90%
	SS	-	-	-	5%	-	20%	30%	25%	100%
P16	TS	-	-	-	10%	50%	-	-	50%	-
	SS	-	-	-	-	-	-	-	-	-
P17	TS	-	-	-	-	-	-	-	20%	20%
	SS	-	-	-	-	-	-	-	5%	-
P18	TS	-	-	-	40%	30%	10%	-	-	-
	SS	-	-	-	25%	20%	-	-	-	-
P19	TS	-	-	-	-	10%	10%	80%	40%	60%
	SS	-	-	-	-	-	-	60%	15%	10%
P20	TS	-	-	-	30%	30%	40%	60%	30%	30%
	SS	-	-	-	30%	15%	-	-	5%	-
MEAN	TS	4%	3%	0%	25%	15%	19%	30%	30%	35%
	SS	1%	1%	1%	10%	7%	10%	14%	16%	22%
SD	TS	7%	4%	0%	22%	22%	21%	29%	26%	31%
	SS	6%	2%	4%	18%	12%	16%	21%	22%	28%

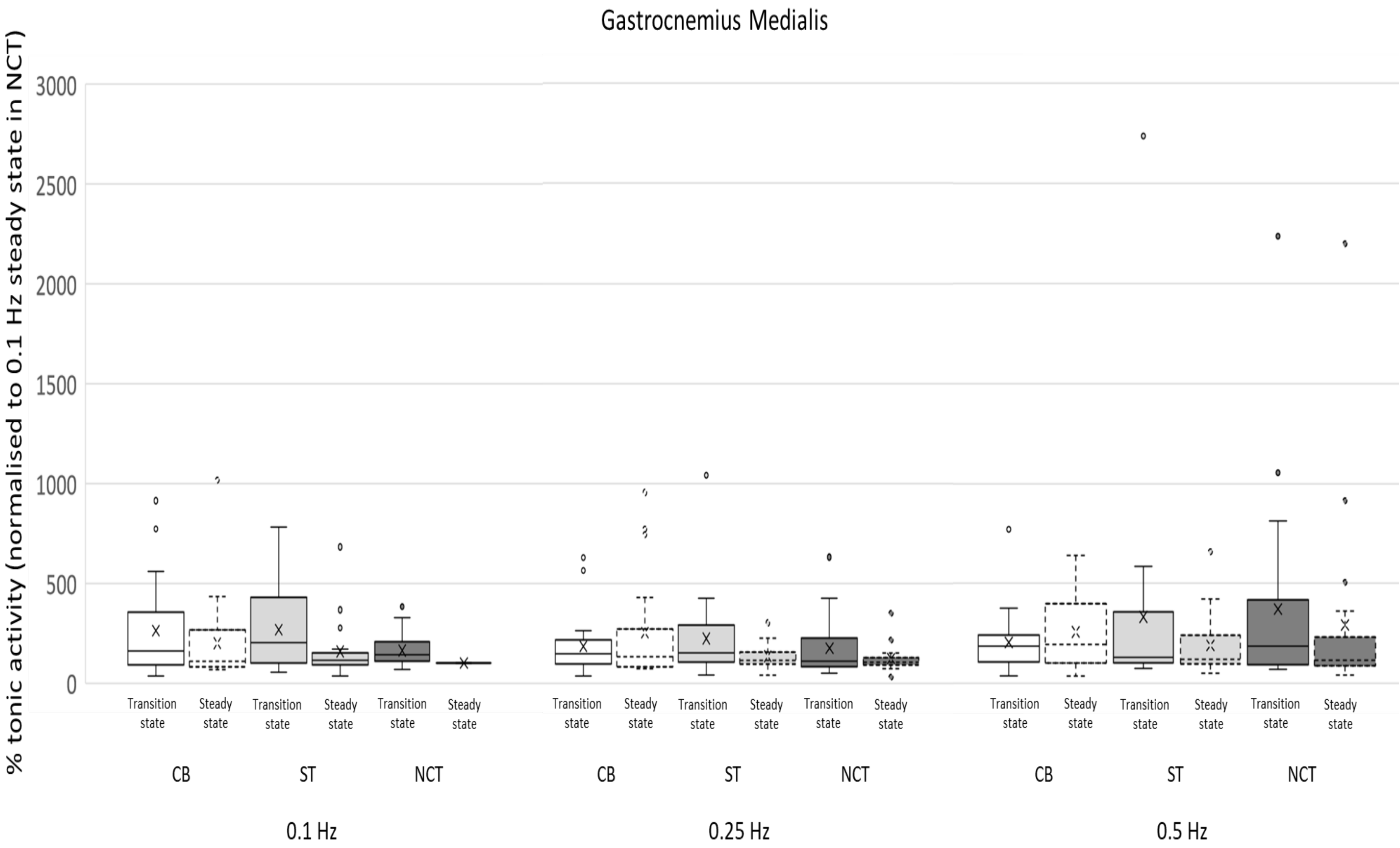
Appendix 2A Percent tonic activity in transition and steady state for each muscle (with outliers).



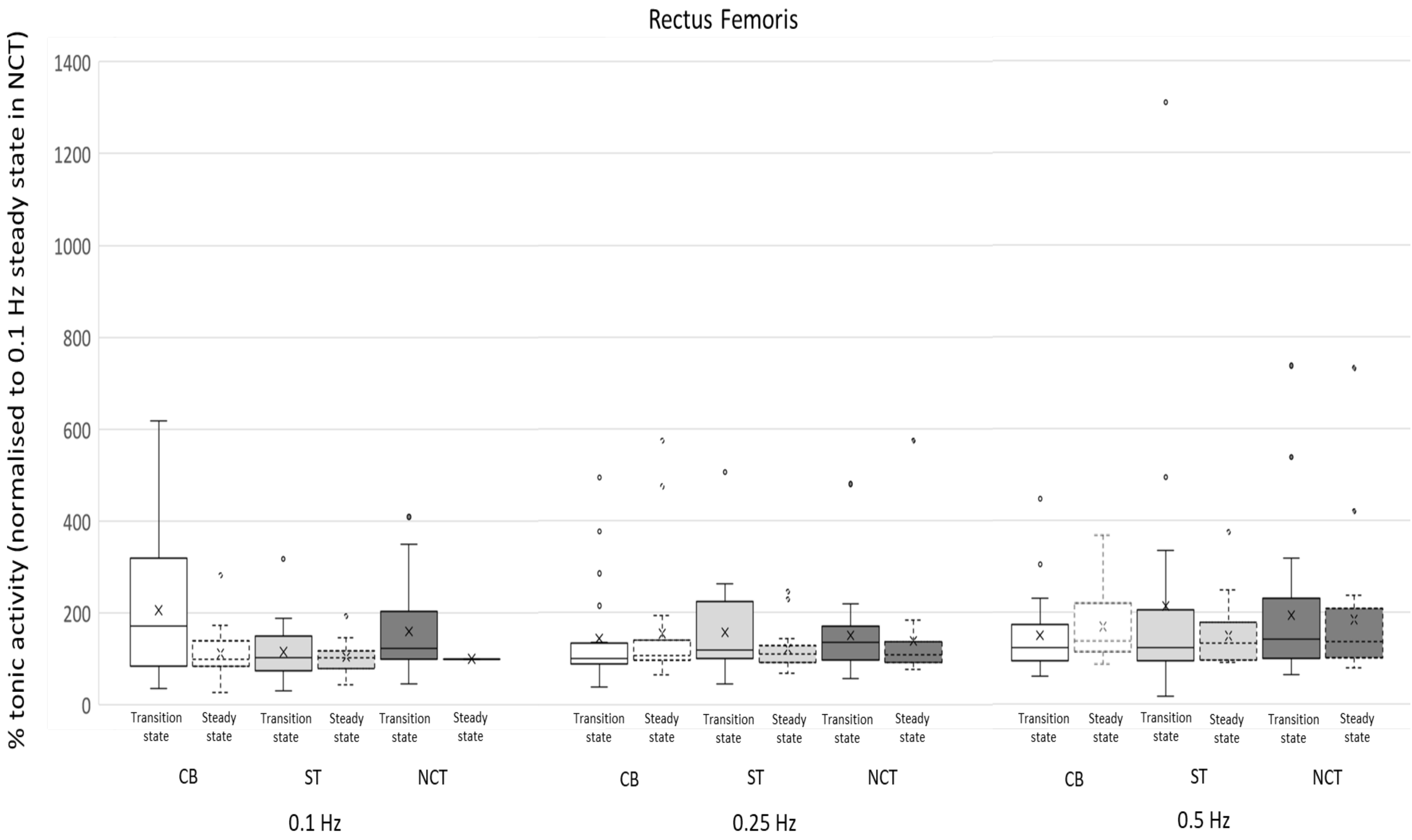
Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Tibialis Anterior as box and whiskers. Dots represent outliers, the x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.



Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Bicep Femoris presented as box and whiskers. Dots represent outliers, the x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.

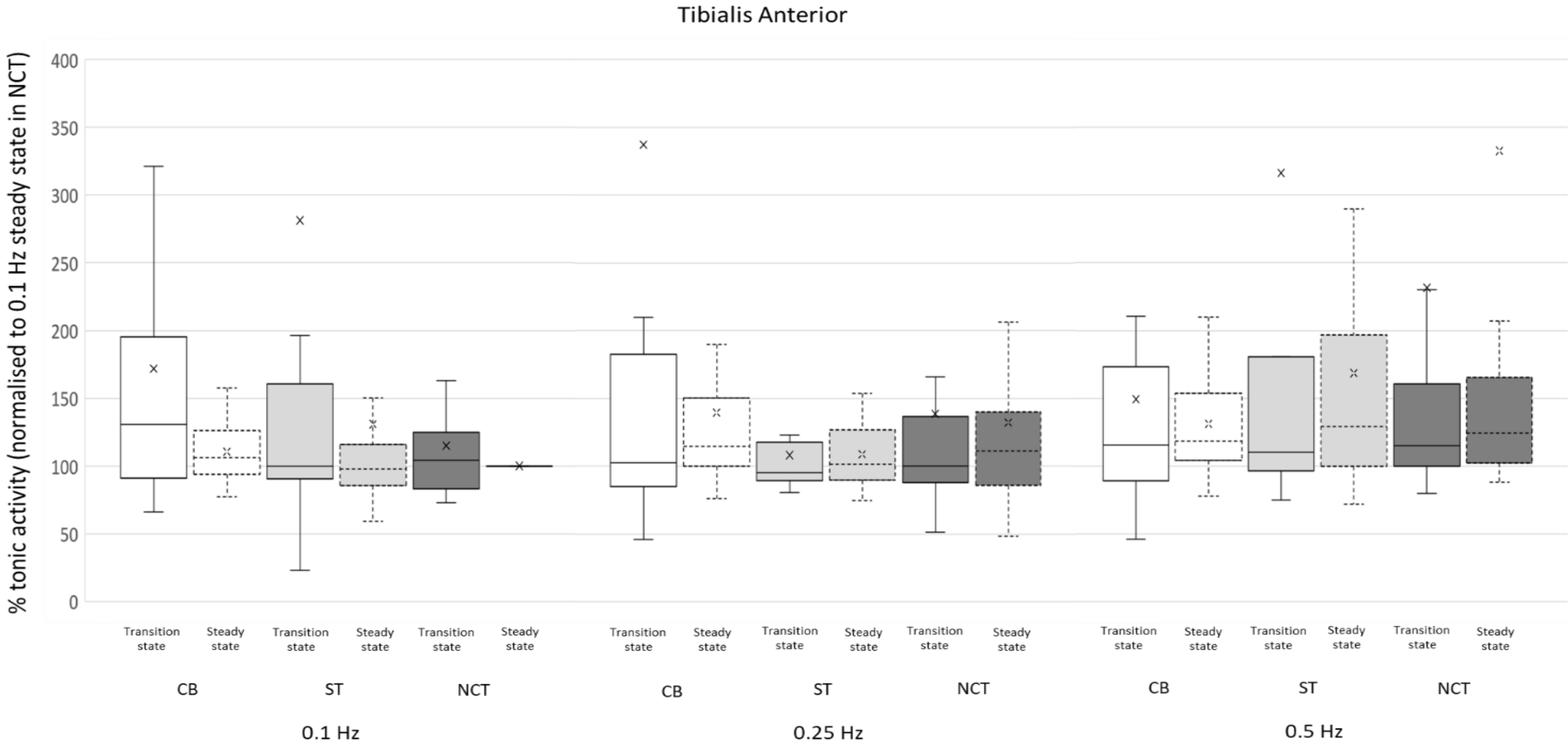


Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Gastrocnemius Medialis presented as box and whiskers. Dots represent outliers, the x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.

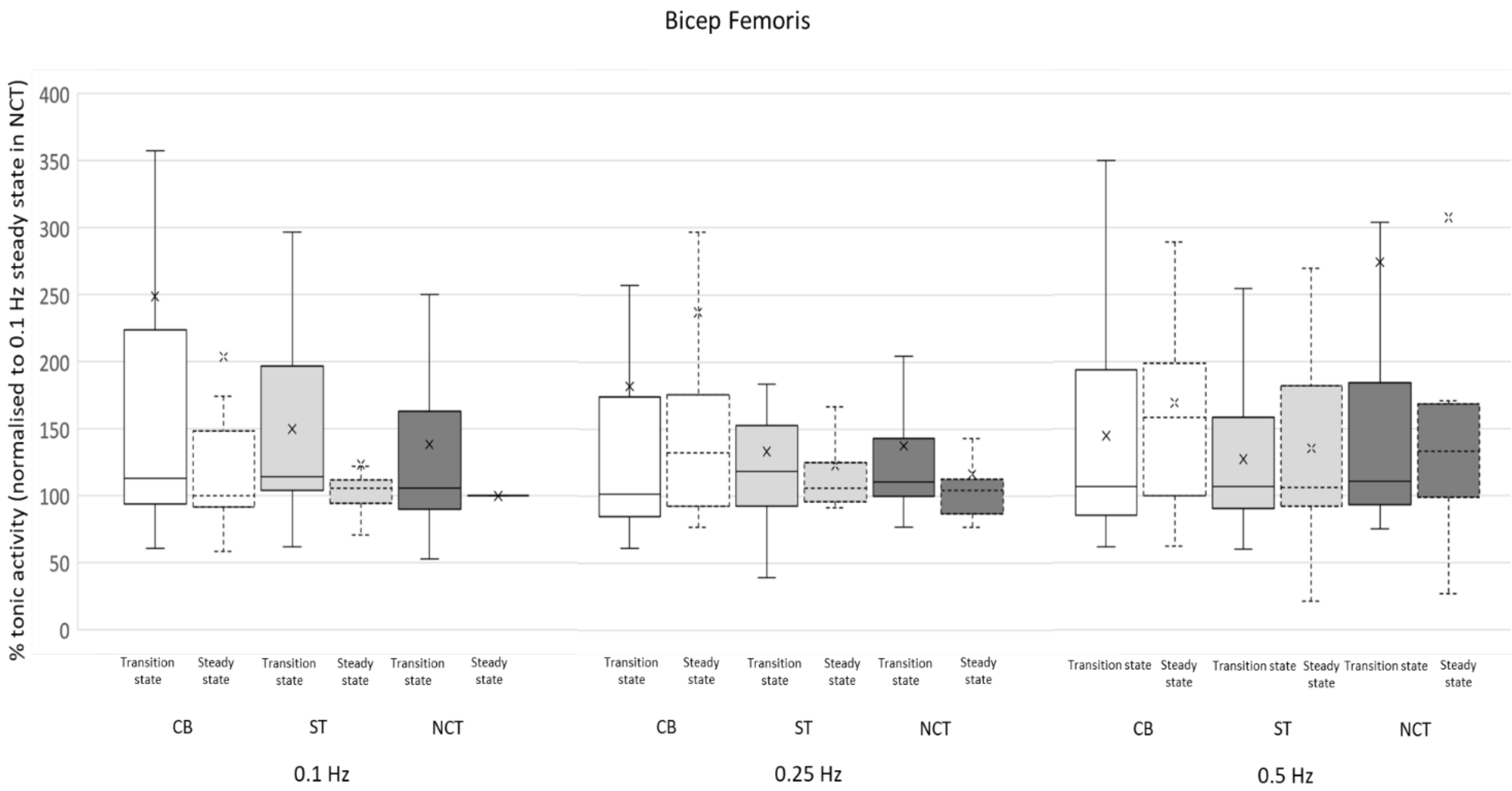


Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Rectus Femoris presented as box and whiskers. Dots represent outliers, the x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.

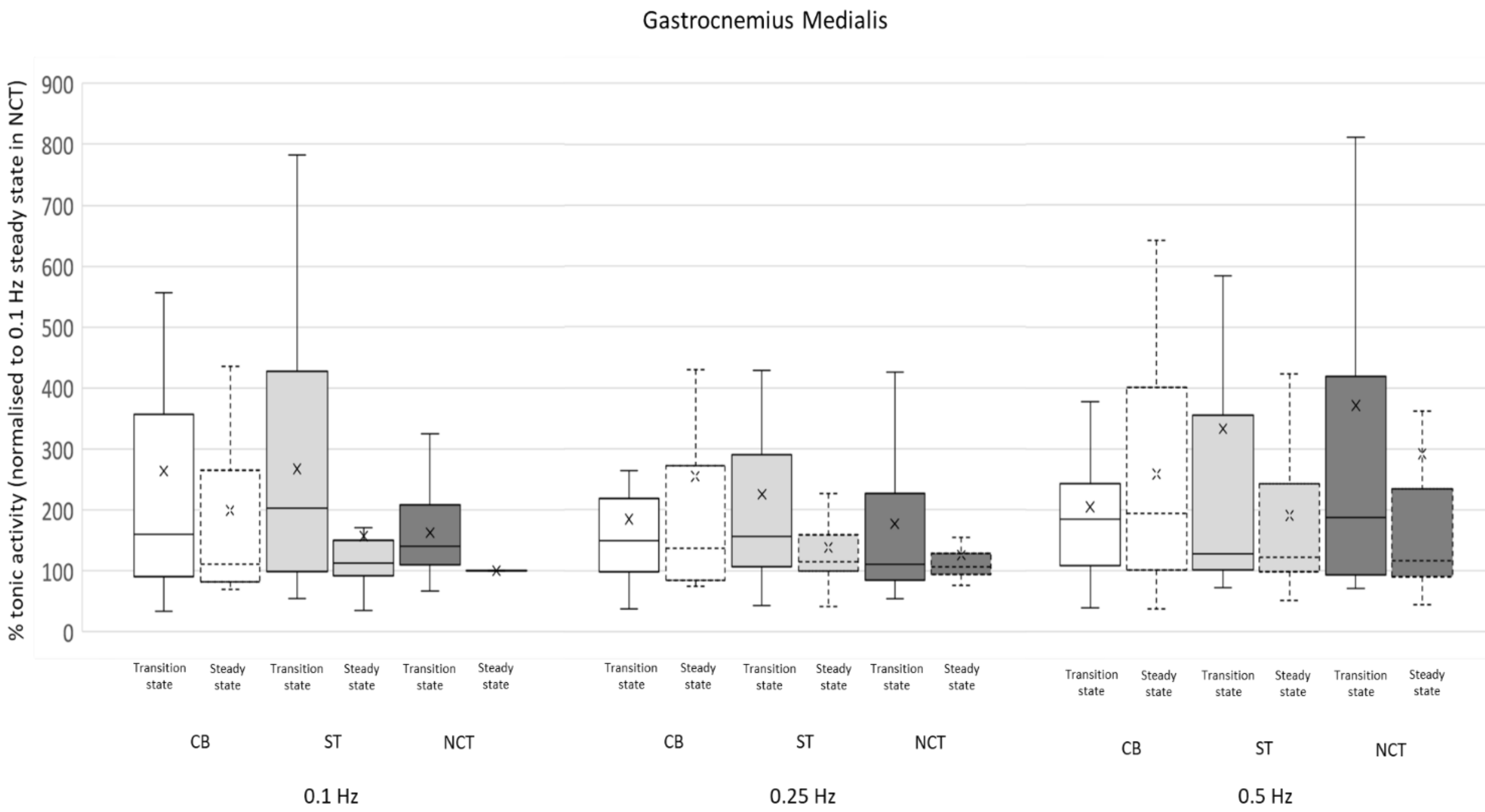
Appendix 2B Percent tonic activity in transition and steady state for each muscle (outliers removed for clarity).



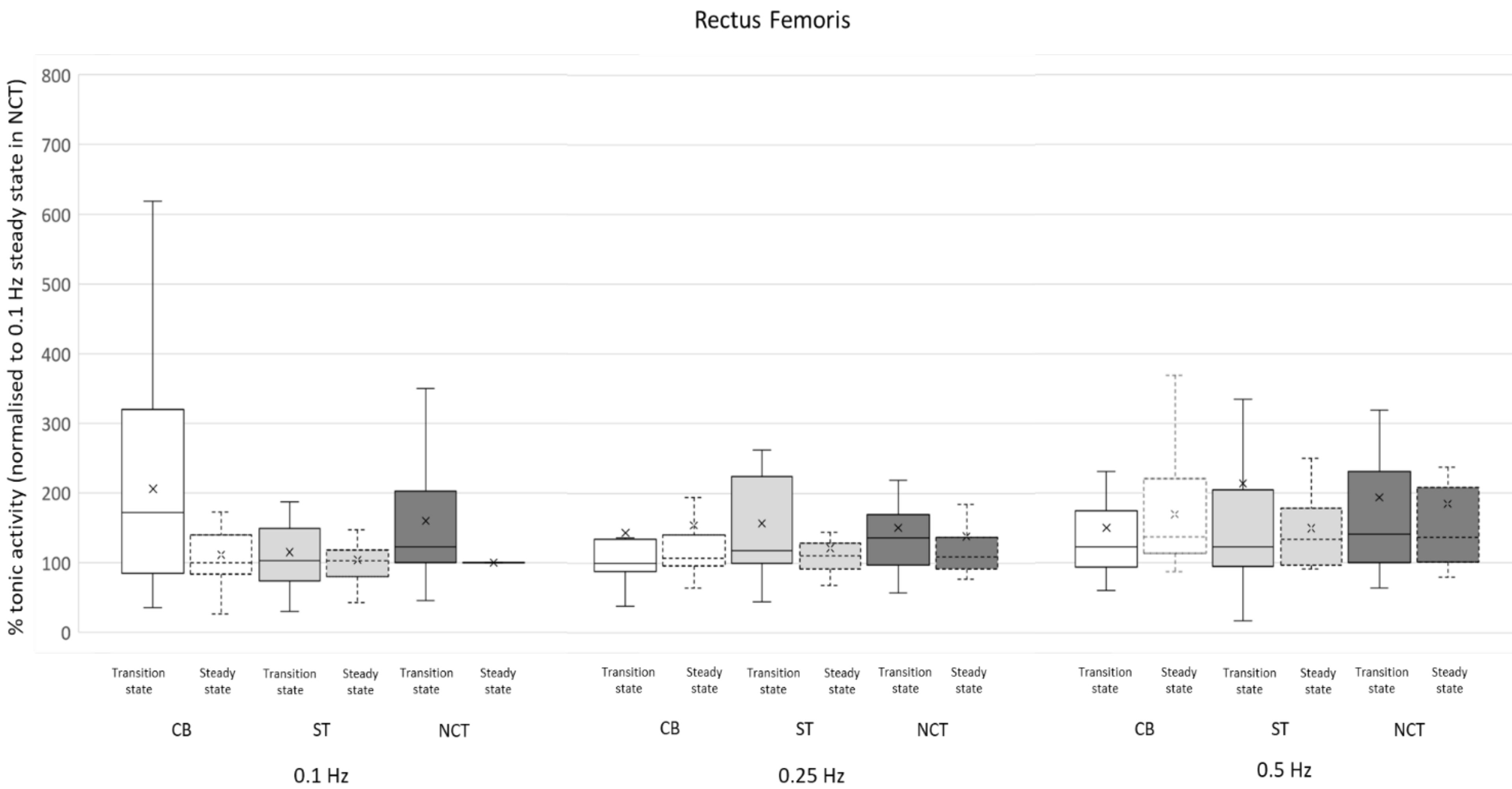
Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Tibialis Anterior presented as box and whiskers with outliers removed. The x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.



Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Bicep Femoris presented as box and whiskers with outliers removed. The x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.



Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Gastrocnemius Medialis presented as box and whiskers with outliers removed. The x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.



Percent tonic activity (normalised to 0.1 Hz steady state in NCT) in transition and steady state for the Rectus Femoris presented as box and whiskers with outliers removed. The x represents the mean, the horizontal line in the box represents the median, while the whiskers represent the minimum and maximum data point.