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Subconcussive impacts to the head have become a growing area of research and concern in the athletic setting. While knowledge on the short- and long-term consequences of concussions has been identified, there is relatively less research on the effects of repetitive subconcussive impacts. Research has shown that neuromotor deficits (i.e., dynamic balance) can be detected acutely after repeatedly heading a soccer ball (a laboratory-based way to induce subconcussive head impacts), but this has typically been done with expensive and non-portable laboratory equipment. However, the AccWalker smartphone application may allow for an objective cost-effective test to examine the effective of repetitive subconcussive exposure. Nonetheless, while cost-effective and portable (e.g., a smartphone app), there is a need for examination of its reliability. Moreover, the extent to which cortical activity is related to dynamic balance control is not well understood. If an association between cortical activity is observed, an increase or decrease in the strength of the association after repeated subconcussive head impacts could be used as an indicator of nervous system impact. These gaps in the literature will be addressed through three specific aims in this dissertation 1) to investigate the reliability of the AccWalker app as a test for neuromotor performance before and after light athletic activity (e.g., kicking a soccer ball); 2) compare EEG spectral power characteristics of dynamic balance across three different AccWalker conditions, and 3) to examine correlations between EEG spectral power characteristics and temporal and spatial kinematic data during a stepping in place task (mTBI Assessment of Readiness Gait Evaluation Test (TARGET)). It was hypothesized that, 1) temporal and spatial characteristics of dynamic balance will not significantly change between pre- and post-soccer kicking activity, 2) EEG power spectral density (PSD) within the

delta and theta frequency bands will increase across the three AccWalker conditions, and 3) EEG PSD within the delta and theta frequency bands will correlate with the temporal and spatial kinematic variables measured using the AccWalker TARGET protocol.

Twenty-four participants were enrolled in this study. Aim 1 used a pre-test/post-test design. Both pre- and post-testing included using the TARGET protocol before and after kicking ten soccer balls. The findings for aim 1 indicated that that the AccWalker TARGET protocol displayed good test-retest reliability with similar data characteristics to previous work. Aim 2 results revealed that EEG PSD measures increased compared to the resting condition. Finally, for aim 3, several significant correlations between the AccWalker spatial metrics within the Delta and Theta frequencies were found.

These findings suggest that postural control assessment can be measured reliably in a pre- to post-test design. This may be important as the AccWalker TARGET protocol may offer a reliable test for changes in neuromotor performance and the body's ability to adapt to "real-life" (or more dynamic) situations. Additionally, this study has expanded on previous literature indicating increased involvement of the frontal-central and central regions of the brain during perturbed balance. Further, this study expands upon the simultaneous use of EEG and balance assessment; specifically, as it is the first study to use a truly dynamic balance task along with a 32-electrode mobile EEG system. This may be important for continued study of not only unaffected balance, but that study of neural changes due to injury or pathological processes.

NEURMOTOR AND ELECTRO CORTICAL ACTIVITY CHARACTERISTICS OF

DYNAMIC POSTURAL CONTROL

by

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Approved by

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DEDICATION

This dissertation is dedicated to all the brains of the world – I now understand why people don’t “get” you.

APPROVAL PAGE

This dissertation written by John Matthew Palazzolo has been approved by the following committee of the Faculty of The Graduate School at The University of North Carolina at Greensboro.

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CHAPTER I: INTRODUCTION

A concussion is a type of traumatic brain injury which can occur when a significant external force directed at the head, resulting in an abrupt change in velocity of the brain resulting in linear and/or rotational movement of the brain within the skull (Centers for Disease Control and Prevention (CDC), 2019). It is estimated that a within the United States, approximately 3.8 million concussions occur each year (Langlois et al., 2006). While considerable research has investigated the effects of acute and long term-effects of concussions on the brain, there is relatively less research on the identification of injury from repetitive subconcussive impacts. A subconcussive impact is a head impact that does not produce the clinical symptoms associated with a concussion (Belanger et al., 2016; Caccese et al., 2018; Mainwaring et al., 2018). negative outcomes. Soccer includes over 265 million active players worldwide and players may “head” a ball up to 300 times per season (Caccese et al., 2018); soccer heading is an offensive and/or defensive maneuver in which a player uses their head to impact a ball to redirect the ball’s path during play. Furthermore, participation in high school and collegiate soccer are at the highest measured (Irick, 2018; NFSHSA, 2017). In response to these rates, there has been increased interest in the investigation of how soccer heading may affect brain health. While average head accelerations during heading may be below a safe threshold (Babbs, 2001), the frequency of these small perturbations may result in acute and/or cumulative damage to white matter structures of the brain (Mainwaring et al., 2018; Tarnutzer et al., 2017).

The assessment and detection of concussive symptomology has received much interest, but less work has investigated on the identification and assessment of repetitive subconcussive impacts. Impaired postural control, cognitive performance, as well as changes in metabolic function of the brain has been previously shown to be affected by repetitive subconcussive exposure (Bailes et al., 2013; Breedlove et al., 2014; Haran et al., 2013; Lipton et al., 2013; Manning et al., 2020; Tarnutzer et al., 2017). Traditional postural control measures and other assessments may either lack required sensitivity to impaired function, can only measure the downstream effects of these changes, and also may not be able to provide for a relatively acute temporal window of effects at the brain-level (up-stream). To close these gaps, recent advances in electroencephalography (EEG) have allowed for the usage of portable (EEG) devices during dynamic whole-body movement. In addition, the advances in smartphone technology have also allowed for the development of mobile assessments of dynamic postural control measurement.

To narrow the current gaps in literature, this study will employ a series of experiments within three manuscripts. The aims and associated hypotheses for each manuscript are presented below:

Manuscript I

Aim 1: To investigate the reliability of the AccWalker app as a test for neuromotor performance before and after light athletic activity (kicking a soccer ball).

- Primary hypothesis #1: Temporal and spatial characteristics of dynamic balance will not significantly change between pre- and post-soccer kicking activity.

Manuscript II

Aim 2: Compare EEG spectral power characteristics of dynamic balance across three different AccWalker conditions.

- Primary hypothesis #2: EEG power spectral density (PSD) within the delta and theta frequency bands will increase across the three AccWalker conditions.

Aim 3: Examine correlations between EEG spectral power characteristics and temporal and spatial kinematic data during a stepping in place task (mTBI Assessment of Readiness Gait Evaluation Test (TARGET)).

- Primary hypothesis #3: EEG PSD within the delta and theta frequency bands will correlate with the temporal and spatial kinematic variables measured using the AccWalker TARGET protocol.

CHAPTER II: LITERATURE REVIEW

Overview

This literature review will discuss the neuromotor systems pertinent to postural control along with the biomechanical systems used to maintain an upright stance. Followed by this, this review will discuss the common postural control assessments used to measure the effectiveness of these systems and postural control strategies. Next this literature review will discuss how postural control assessments can measure the effects of altered information due to possible system compromise following concussion and/or subconcussive repetitive head impacts. Finally, a discussion will be presented on the use of electroencephalography in the study of postural control and include a discussion on the common methods used with electroencephalography research.

Neuromotor Systems

Neuromotor control reflects the integration the sensory systems of the central nervous system (CNS) and peripheral neuromuscular systems. Balance (a neuromotor task) is the ability to maintain an upright posture of one's body. Balance is accomplished by maintaining one's center of mass within one's base of support. However, whether in a static or dynamic state, the upright posture is innately unstable as any small deviation of the center of mass of a body possess the ability to produce sufficient force by the acceleration of the body due to gravity to displace the body's center of mass away from the base of support, resulting in instability and thusly a fall (unless additional body maneuvers are quickly incorporated). Hence, constant monitoring of the position of the body within space is needed by three main sensory systems; the somatosensory (proprioceptive), visual, and vestibular systems in a fluctuating manner (Peterka, 2018). The proprioceptive, visual, and vestibular systems are organized throughout the head and can work in concert with each other to meet five tasks, resist gravity to maintain and upright posture, maintain the body's center of mass within its base of support, provide for postural stability of the task, provide for acceptable foot-ground clearance, and control for the accelerations of the head to stabilize visual and vestibular systems (Winter, 1989). Using balance related tasks allows for the assessment of the current capabilities and proportion of each of these systems during said task and may allow for the assessment of potential altered ability of these systems and other central nervous systems, subsequent to a concussion of repetitive subconcussive head impacts (Guskiewicz et al., 2001; Rhea et al., 2017).

The proprioceptive system allows for knowledge of the location, orientation, velocity of movement of the body and its limbs in relationship to the body's base of support and is transmitted to the nervous system via multiple proprioceptive and cutaneous mechanisms (MacKinnon, 2018). One of these are the muscle spindles embedded within muscle tissue containing three types of intrafusal fibers; nuclear chain, static nuclear bag, and dynamic nuclear bag fibers, which function as receptors which can signal both the length (re. stretch) and lengthening velocity of a muscle (Purves et al., 2012). These fibers are innervated by two sensory fibers, Group 1a and Group II afferents. Group Ia innervate all of the three fibers and have the ability to relay information about muscle the length of the muscle as well as its and stretch velocity and are more sensitive to minor changes in muscle position (MacKinnon, 2018). While Group II fibers do not innervate the dynamic nuclear bag fibers, thusly providing information on muscle length only (Purves et al., 2012). Furthermore, due to differences in myelin composition and axon diameters, with Group Ia afferents being greater in both respects, Ia fibers display greater conduction velocities than their Group II counterparts (Purves et al., 2012). Furthermore, the target destinations for these afferents are different as well. Group Ia afferents perform excitatory monosynaptic connections with the α-motoneurons (α-MN) of an antagonist muscle resulting in short-latency reflex (stretch-reflex), providing quick countermovement to posture disturbances (MacKinnon, 2018). While Group II afferents slower conducting signal terminate upon an interneuron that can provide for either an excitatory or inhibitory input to the α-MNs of the ventral horn of the spinal cord (with inhibitory interneurons projecting both ipsi- and contralaterally), providing for the sensory information of limb position and error feedback for the long-latency modulation of postural correction (MacKinnon, 2018). With these systems working in concert, perception of position and required rapid changes in posture (posterior and anterior sway), as well as the regulation in movement rhythm (stepping-in-place), performance can be achieved. Additionally, the Golgi tendon organ (GTO) can provide information of the level of tension that may be applied to the tendon of a muscle (Purves et al., 2012). The GTO are located in series at the junction between the muscle and the tendon and is innervated by a singular Group Ib afferent. A signal is propagated during muscle contraction, causing the straightening of the collagen fibers of the tendon and compressions the GTO (Purves et al., 2012). The Ib afferents then bifurcate (spilt in two) entering the spinal cord. The key role of the GTO is to monitor produced muscle force via tendon length and tension (MacKinnon, 2018). By monitoring these variables, the signals propagated via the Group 1b afferents and information on the load of a limb may play an important role in postural control and modulation as well as postural control during rhythmic movement through the transition from flexion to extension at a lower limb joint (Kistemaker et al., 2013; MacKinnon, 2018). Additionally, the proprioceptive system is also informed by GTOs in the joint ligaments Ruffini endings and pacinifrom-like corpuscles within the joint capsule (MacKinnon, 2018). While these receptors may contribute little to proprioceptive feedback in the normal range of motion (RoM) of a joint, at extreme ranges, increased activity of Group II afferents have been shown to excite premotor interneurons resulting in suppressed activity in the agonist musculature (MacKinnon, 2018). Finally, cutaneous receptors (especially in the feet) can provide information on the amount and displacement of pressure beneath the foot (Purves et al., 2012). The information provided by these systems may provide information for the coordination of rhythmic movements. For example, when a leg is unweighted, the activation of the cutaneous receptors within the sole of the foot can cause a reflexive suppression of the leg extensors and excitation of the leg flexors (MacKinnon, 2018; Purves et al., 2012).

The visual system collects and transmits information of a body's place in space through static and dynamic environmental information of the size, distance, and detail of the space surrounding the body and/or objects within that space. Visual information can be used the planning and execution of postural corrections and reactions. The change in posture that are within the visual field are first gathered at the retina and delivered to the visual cortex located within the occipital lobe via the optic nerve (Purves et al., 2012). The output from the visual cortex is then through two functionally different pathways, the posterior and anterior stream (Goodale & Milner, 1992). The anterior stream acts to process the shape(s) of an object and environment and is used to generate plans for future actions based upon this perceived space (Goodale, 1996). The posterior stream conveys information to be used for the control of an action in real-time by changing the aforementioned information and essentially overlaying a frame of reference for the limbs and body so that actions may be taken (Goodale, 1996). When the visual field becomes perturbed, quick correction of the body can be from a planned set of actions via the superior colliculus and vestibulo-ocular and optokinetic movements. Upper motor neurons in both the superior colliculus and frontal eye fields, each of which contain a topographical map of eye movement vectors, discharge immediately prior to saccades. Activation of a particular site in the superior colliculus or in the frontal eye field produces saccadic eye movements in a specified direction and for a specified distance (Purves et al., 2012). Vestibulo-ocular and optokinetic movements are complementary to each other and function to stabilize gaze by countering the movement of the head (when the eyes are moving opposite to the head). While optokinetic movement is both smooth and saccadic, and if head movement is slowed the optokinetic movement will compensate by producing movements of the eye that maintain foveation of the object (Purves et al., 2012).

Finally, the vestibular system which can quickly sense both linear and rotation head movements is used to assist in the stabilization of the head and gaze during motion, as well as provide the sense of motion that can be used to initiate postural reflexes to maintain the upright stance. Information for the vestibular system using the semicircular canals and otolith organs (Purves et al., 2012). Three semicircular canals are located within the temporal bone. Each of these has its base at the ampulla, which contain the crista housing hair cells which detect rotational accelerations of the head. These hair cells extend out of the crista into a gelatinous mass (capula) covering the width of the ampulla forming an enclosed space in which circulation is minimized (Purves et al., 2012). The semicircular canals detect rotational head movement. Since all of the hair cells within the semicircular canal are orientated in the same direction, if the head were to rotate, the capula becomes distorted (opposite of the direction of movement). Distortion of the capula causes a complimentary movement of the hair cells located within the crista, and the movement of the hair cells depolarizes the cells (increase in potassium ions); generating nerve impulses to the 8th carinal nerve (Purves et al., 2012). The two otolith organs (utricle and saccule) which lie against the inner ear between the semicircular canals and the cochlea and are orientated at 90 degrees to each other contain patches of hair with the terminal end of these hairs having small calcium carbonate crystals (otoconia). These hair cells are embedded within each organ in the macula (Purves et al., 2012). The otolith organs detect tilting and translational movement of the head. Detection of tilting and translational movement is accomplished by the constant activity of the hair cell's nerve fibers within the macula. When the head is displaced, the firing rate of the associated axons is increased or decreased depending on the direction of head movement. It is increased on the side ipsilateral to the direction of movement and decreased on the contra lateral side (Purves et al., 2012). The vestibular-cerebellar circuit is important in the integration and control in vestibular activity. Specifically, the vestibular-ocular reflex, which is important for creating eye movements that are opposite of head movements, resulting in a fixed gaze upon on object while the head is moving. The vestibular-ocular reflex is important for sensory integration while the body or head is in motion, as in walking. This integration is also important as it may assist in the determination of active verse passive movement of the head. The integration of sensory motor responses also assists in the coordination of axial musculature and coordination of posture. The vestibular-cortical circuit is composed of superior and lateral vestibular nuclei to the posterior thalamus which are then integrated into the somatosensory and motor cortex. The integration can assist in integration of vestibular and visual stimuli. Allowing for a better sense of body position and movement when visual, vestibular, proprioceptive information is integrated. This circuit also assists in the control of proximal musculature and posture via descending pathways of the vestibulospinal tract and cortico-vestibulospinal tracts (MacKinnon, 2018).

Postural Control Assessments

Balance assessment (postural control) assessment is considered a best practice in the management and detection of concussion, repetitive head trauma, and other neurological conditions (Guskiewicz et al., 1996; Riemann & Guskiewicz, 2000). There are a variety of balance assessments that can be employed to assess neuromotor performance and these assessments can be separated by either static or dynamic balance assessments, or objective or subjective assessments. The choice of an assessment should be guided by the demands resources available as dictated by the environmental setting and research question.

An objective measure of postural control is the assessment of CoP variables gathered through the use of a force plate (Guskiewicz et al., 2001). One such CoP variable is CoP displacement. As an individual maintains an upright posture whilst standing on a force plate (resisting their body weight as a function of the acceleration of gravity), the proprioceptive, vestibular, and visual sensory systems work in concert to maintain this stance through the use of active postural corrections, leading to a displacement of the CoP (MacKinnon, 2018). While CoP displacement has been shown to be a valid and reliable measure for altered sensory ability resulting from concussion (Guskiewicz et al., 2001). It has been argued that CoP velocity (the rate at which CoP changes direction and speed of displacement) may be more sensitive under dynamic conditions in participants with concussion and other neurological conditions (Parker et al., 2006; Prieto et al., 1996; Roeing et al., 2016). The current gold-standard for postural control assessment is the Sensory Organization Test (SOT) (Guskiewicz et al., 2001). The SOT possesses the ability to individually alter each sensory system, by altering the external information provided to these systems across six conditions, repeated three times over 20s each (Guskiewicz et al., 2001). The SOT represents the use of a combination of balance assessments working in conjunction with a force plate under dynamic conditions (Guskiewicz et al., 2001), Each of the six conditions a combination of fixed and dynamic modalities; the first condition consists of the participant standing on a fixed force platform with their eyes open, the second condition consists of a fixed platform but with the eyes closed, the third condition requires a participant to stand again on a fixed platform with their eyes open but the visual system is perturbed by a sway-referenced visual surrounding, the fourth is similar to the third but while the sway referenced visual surrounding is fixed, the force platform itself is sway referenced, the fifth condition is similar to the fourth, but the participant will be in a vision deprived state (eyes closed), finally, the sixth condition requires the participant to stand with their eyes open, but whilst the force platform and visual field is sway-referenced (Cohen et al., 1993). During each of these conditions postural sway dynamics are collected and through a combination of these values and conditions, the proprioceptive, visual, and vestibular system performance is measured (Jones et al., 2011).

However, the SOT may be limited by factors such as cost and time of testing, as well as the requirement of designated laboratory space for assessment (Schmidt et al., 2012). In response to these limitations, alternative assessments have been developed. Possibly the most common subjective static test is the Balance Error Scoring System (BESS). The BESS assessment requires participants to maintain three different postural stances over two 20s trials (eyes closed), with or without (modified BESS) foam pad support (Guskiewicz et al., 2001). Yet, while the BESS test may allow for an alternative to traditional laboratory measures for the identification of concussions, due to its subjective nature it demonstrates low sensitivity and rater reliability (Guskiewicz et al., 2001; Guskiewicz & Broglio, 2015; Murray et al., 2014; Simon et al., 2018). As a response to the limitations of the BESS test, there has been research investigating the use of virtual reality (VR) and virtual environments (VE) as a mimic to the SOT protocol (Wittstein et al., 2020). Prior research has been performed to assess the validity of a VE in simulating the BESS balance test for concussion assessment. Simon et. al., (2018) has provided some evidence to suggest that a VE may increase the sensitivity of the BESS. In a study involving 28 adults, researchers performed a within-subjects comparison study of the traditional BESS test to a VE representation of a rollercoaster ride, presented by an LG V10 smartphone via Google Cardboard. It was determined that the VR BESS test significantly increased both total errors and CoP velocity across all conditions of the BESS test (Simon et al., 2018). These findings suggest that VR can successfully be used to implement balance assessments (Simon et al., 2018). Again however, while the CoP measurement were measured objectively through a force plate, the BESS errors were still subject to subjective rater reliability (Murray et al., 2014). Nonetheless, VR may be useful in the mimicking of the already validated objective SOT (Teel & Slobounov, 2015). Work by Wittstein, et al., in 2020, created a VE representation of the six conditions of the traditional NeuroCom SOT (with the exception of a sway reference force platform, this was replaced with a foam cushion support overlaid on top of a force plate), and compare the condition of the VE to the traditional SOT. The results of this investigation found that several of the VE conditioned corelated well with conditions of the traditional SOT (Conditions 1, 2, and 3) (Wittstein et al., 2020). However, the somatosensory conditions (4 and 5) displayed reduced intraclass correlational coefficients compared to the traditional SOT, suggesting that a foam pad may not possess the ability to properly alter proprioceptive information compared to a sway-referenced support (Wittstein et al., 2020). Additionally, the Clinical Test of Sensory Interaction and Balance (CTSIB) (Cohen et al., 1993). The CTSIB is a subjective assessment of balance with conditions similar to the SOT (Guskiewicz, 2001). With both vision and physical support altered, but through the use of a visual dome obstruction and foam support pad, respectively (Cohen et al., 1993). While validity and reliability of this assessment is good, it may be limited in assessing any detriments to sensory systems beyond the vestibular system (Cohen et al., 1993). However, these two aforementioned assessments are of a more static nature and may not impose demands sufficient enough to detect small changes in postural control (Rhea et al., 2017). The Community Balance and Mobility Scale (CB&M) offers a dynamic subjective assessment of balance through the use of a 13-task assessment (e.g., timed walking, carrying, and stepping tasks). While subjective in nature (the assessor may score in real-time or via video-recorded sessions), it has been shown to be a valid and reliable measure for neuromotor performance of individuals with and without balance abnormalities (Howe et al., 2006).

A common separation of assessments is how the task is performed, notably is the assessment a measure of static or dynamic balance. A static balance test, until recently required the use of high-cost force plates requiring designated laboratory space (Schmidt et al., 2012). However, recent technologies have come to market that provide a low-cost alternative and portability (Goble et al., 2016; Patterson et al., 2014). Static balance tests are classified as such because, during each assessment the participant is required to remain as still as possible in the testing position. To operationally define static balance, either CoP displacement or other forms of displacement are determined from the initial starting point (position at start) and are continually recorded throughout the test. Contrary to static balance assessments, dynamic balance assessments may allow for increased resolution of alterations of the three sensory systems (Goble et al., 2016; Guskiewicz et al., 2001; Kuznetsov et al., 2018; Rhea et al., 2017). Through the use of a standard movement protocol (e.g., stepping-in-place) over several conditions (eyes open, eyes closed, or complimentary head movement whilst stepping-in-place), not only can smaller deficits in balance be assessed, but alterations in the three sensory systems may be elicited (Kuznetsov et al., 2018; Rhea et al., 2017). Common variable data from dynamic tests include RoM and rhythm consistency (or variability of that rhythm), and the differences in this data pre- and post- either a neurological injury or therapeutic modality may suggest alterations in the three sensory systems (Rhea et al., 2017).

The choice between the use of static, dynamic, subjective, and objective balance assessments should be determined by the needs of the research question or reason for assessment (Rhea et al., 2018). Further, while subjective tests may allow for a more economical assessment, they may be subject to lower sensitivity and reliability (Murray et al., 2014). Additionally, while the current market does offer objective static balance tests, the relative simplicity of the task (requiring less vestibular input), may limit the ability to detect small changes in neuromotor control, dynamic tests may be able to detect (Baloh et al., 1994; Rhea et al., 2017). If one requires the assessment of the body's ability to adapt to more "real-life" situations and perturbations of the three sensory systems, an objective and dynamic assessment should be selected (Rhea, et. al., 2018).

Head Trauma and Postural Control

The maintenance of balance is regulated by muscular forces produced from interpretations of the information and feedback from the three aforementioned systems (Assländer & Peterka, 2014). However, the contribution of each individual system may be altered according to a specific perturbations or possible pathological changes to one or a combination of these systems. A particularly variable system may provide incorrect information to another, resulting in decreased balance performance (Peterka, 2018). For instance, a neurodegenerative condition may affect the proprioceptive system (Cattaneo et al., 2016).

Furthermore, vestibular system performance is significantly altered post a concussion (Guskiewicz et al., 2001). In addition, it may be important to investigate not only the extent to which an impaired system may affect balance performance, but also whether with training these systems could be "re-weighted" following a rehabilitative measure.

Through the use of balance assessment, certain aspects, or detriments of neuromotor control may be studied and the concept of a re-weighing of systems can be assessed. In 2011, Peterka and colleagues studied sensory re-weighing in 11 participants with unilateral vestibular loss (UVL), a condition resulting in functional compensation in balance and gait (Peterka et al., 2011). Postural control was assessed using an intervention in which a tiltable support surface was rotated at four different amplitudes (participants remained in an "eyes-closed" condition, the visual system would be absent); with the hypothesis being that in the absence or altered ability to employ the vestibular system, the weight of contribution of the proprioception would be increased compared to controls. Data from the Peterka et al., (2011) study indicated a significant difference in sway response between controls and UVL participants, with UVL participants demonstrating increased sway relative to the control sample. This data suggested that in an altered ability to employ the vestibular system, other systems (demonstrably, the proprioception system) express higher contribution to postural control (Peterka et al., 2011)

Additionally, previous research has demonstrated a dynamic re-weighing of these systems following balance training modalities. In 2015, Cone and colleagues studied how an exer-gaming modality (Wii Fit Plus), may result in a re-weighing of the vestibular system as measured via the fifth condition of the NeuroCom SOT. The exer-gaming modality consisted of two to four 30 – 45 min/day exer-gaming sessions over the course of six weeks, with SOT testing before and after the completed intervention (Cone et al., 2015). At the end of the six-week intervention endpoint excursion (initial center of pressure displacement (COP (cm) as a percentage of maximal CoP displacement), movement velocity (CoP velocity °/s), and response time (amount of time (s) between initial signal and movement to a specified target), all had significantly improved; suggesting that a re-weighing in favor of the vestibular system can be achieved as a practice effect of balance training (Cone et al., 2015).

A concussion is a form of a traumatic brain injury which can be caused by a sufficient magnitude of kinetic energy transferred through the skull and altering brain structure (e.g., shearing of axons, swelling, and even neuronal death) (Srivastava & Cox, 2018). Additionally, perhaps secondary to these alterations to metabolism within the brain, and blood-oxygen delivery may also be diminished (Werner & Engelhard, 2007). These alterations and processes can also affect the three sensory systems of the nervous system, resulting in impaired information transmission, resulting in decrease neuromotor control (Srivastava & Cox, 2018). A subconcussive impact is a head impact that does not produce the clinical symptoms associated with a concussion (they are considered to be sub-clinical) (Belanger et al., 2016; Caccese et al., 2018; Mainwaring et al., 2018). While the subconcussive impacts may not be considered as injurious as that of concussions (Belanger et al., 2016; Mainwaring et al., 2018), repetitive exposure to subconcussive impacts has been shown to be associated with both acute (Di Virgilio et al., 2016; Rhea et al., 2017) and chronic negative outcomes in neuromotor and neurocognitive ability (Bailes et al., 2013; Breedlove et al., 2014; Downs & Abwender, 2002; Manning et al., 2020). Unsurprisingly, there is evidence that individuals who have had repetitive subconcussive head trauma have worse outcomes, than individuals without impact exposure. (Di Virgilio et al., 2016; Hwang et al., 2017; Koerte et al., 2015; Rhea et al., 2017). A setting in which subconcussive exposure may be common is within the military setting. Blast exposure (from demolition explosives and shoulder-mounted-weaponry) may allow for the transit of force waves through the skull of the head and altering brain structure. Specifically, these waves may have the ability to affect protein shape and integrity, thereby changing the properties of neuronal cell elasticity and strength possibly leading to axonal loss (Dennis & Kochanek, 2007). Research conducted by Rhea et al. (2017) investigated neuromotor function across several times points following blast exposure in 41 active U.S. navy personnel who were present for heavy weapons training using a smartphone application (AccWalker). The AccWalker program was used with an Android-based smartphone The Android device is placed on the lateral thigh of the dominant leg via Velcro and the app is activated while the participant steps in place to their natural cadence. The app records the orientation of the participant's thigh during the stepping task. Spatial and temporal characteristics of the thigh orientation are quantified to examine dynamic balance control (Rhea et al., 2017). Differences in thigh range of motion (RoM), stride time (as well as the standard deviations of these two measures), were found for up to 72 hours post-blast exposure (Rhea et al., 2017).

Additionally, it may be important to consider that the concept of concussive and subconcussive trauma to the head and subsequent neuromotor changes may have a sex-bias. The risk of concussion and severity of symptoms has traditionally been focused on the male sex and male-dominant sports. With the increase in female sport participation, epidemiological data has suggested that female athletes (at both the high school and collegiate levels) are a significantly greater risk of concussion relative to their male counterparts (Covassin et al., 2003; Gessel et al., 2009). A cohort study performed by Covassin et al., in 2003 of 14,591 reported injuries sustained during competition in collegiate sports, found a significantly greater incidence of reported concussions (percentage of total injuries) in females (9.5%) compared to the incidence of concussions in males (6.4%) (interestingly, the greatest density ratio of reported concussion in both sexes was soccer) (Covassin et al., 2003). Additionally, an epidemiological study performed by Gessel et al., in 2009 suggested a similar discrepancy between reported concussion between sexes with again female concussion rates being higher than their male counterparts (Gessel et al., 2009).

Females have also been shown to display poorer post-injury symptomology and outcomes (Baker et al., 2016; Broshek et al., 2005; Chiang Colvin et al., 2009; Covassin et al., 2007, 2012, 2013; J. H. Miller et al., 2016; Zuckerman et al., 2014). Several studies have identified significant differences in neurocognitive and neuropsychological performance between sexes post-concussion. In 2005, Broshek and colleagues provided evidence that return to play decisions and management of concussions should consider sex as a factor. After assessing baseline reaction times in 155 female and male athletes (pre- and post- a diagnosed concussion), it was found that not only were females at 1.7 times more at risk of cognitive impairment from a concussion, but also displayed significantly slower simple and complex reaction times (measured via the Concussion Resolution Index) (Broshek et al., 2005). Further, several prior studies have found poorer post-concussive outcomes in female athletes (compared to male athletes) in the ImPACT battery for up to eight days post-concussion (Baker et al., 2016; Chiang Colvin et al., 2009; Covassin et al., 2007, 2012; Covassin, Elbin, Bleecker, et al., 2013; Zuckerman et al., 2014).

Beyond neurocognitive and neuropsychological assessment differences, postural control differences are also seen between sexes. Postural control assessment is a main variable in return-to-play decisions and can also provide information as to altered neuromotor functioning (Goble et al., 2019; Guskiewicz, 2001). Sex differences in postural control at baseline measures have been previously seen; generally, with females outperforming males (Paniccia et al., 2018). Some prior research has attributed to these differences to better sensory integration in females (Halonen, 1986; Halonen et al., 1986; Paniccia et al., 2018). There is also data to suggest that due to the differing rate of development during puberty, females may outperform in balance measures compared to males (Goble et al., 2019), but recent work may diminish this theory when development is considered (Palazzolo et al., 2019). An additional thought may be that differing concentrations of differing muscle fiber types may allow for better postural control and reactionary movements see in females, especially type-II non-oxidative fibers (A. I. Miller et al., 2015). Yet, previous research is limited in accounting for significant differences in fiber types count and size between male and female human, and even less when controlled to overall body size (Haizlip et al., 2015). Even so, this may not explain the differences in symptom severity and prognostic outcomes seen between the two sexes (Covassin et al., 2003, 2012). In fact, some evidence suggests males may perform significantly better in postural control measures post-concussion (Covassin et al., 2012), which is in contrast to non-injured states in which the opposite is seen (Paniccia et al., 2018). A study of 222 high school and collegiate athletes supports the premise of age and sex being integral to postural control performance after concussion (Covassin et al., 2012). Notwithstanding the results of this study were mixed, with high school male athletes displaying poorer BESS scores than their female counterparts, and the opposite within the college-aged groups (Covassin et al., 2012). The BESS test does possess inherent limitations, notably low sensitivity for identification of concussion and low interrater reliability (Murray et al., 2014). Due to these limitations, a more dynamic postural control assessment may be warranted (especially for sub-concussive exposure) that may also include vestibular system assessment (Sufrinko et al., 2017). Along with a more dynamic postural control assessment the Vestibular/Ocular Motor Screening may allow for more accurate assessment in disturbances in neuromotor control, specifically, the assessment of the vestibulo-ocular reflex (the mechanism of producing countermovement of the eyes during head rotation (Purves et al., 2012) (Sufrinko et al., 2017). In a study involving 64 participants (aged 9-18 years, 28 females and 36 males), the ImPACT battery, Post-concussion, Symptom Scale (PCSS), and the BESS test were used to assess post-concussive symptomology with 21-days post-exposure. While no sex differences were seen for either BESS or ImPACT measures, females did exhibit significantly higher PCSS and vestibular-ocular reflex scores.

Electroencephalography

The identification of the neural mechanisms in human movement is important in the study of postural control (balance). Concussions and repetitive head impacts represent a possible injury to the proprioceptive, visual, and vestibular sensory systems of the CNS, thereby altering neuromotor control. A feedforward and input from these systems is important in the maintenance of balance (MacKinnon, 2018; Peterka, 2018). Additionally, these systems are integral in the monitoring and input needed for the coordination with other neural structures of latency dependent movements (e.g., stepping) (Jacobs & Horak, 2007). While the measurement of the downstream effects of altered sensory systems can be seen during neuromotor tasks (e.g., balance assessment); until recently, the study of the central nervous system simultaneously during movement was problematic, but recent advancement in technology have allowed for the study of cortical activity during dynamic postural tasks possible through the use of electroencephalography (EEG) (Sipp et al., 2013). When compared to other techniques such as functional magnetic resonance imaging (fMRI) and magnetoencephalography (MEG), EEG represents a significantly smaller economical and space intensive cost (Freeman & Quiroga, 2013). Additionally, due to the relatively slow pace of blood hemodynamics, EEG offers exceptionally higher temporal resolution to fMRI techniques, along with greater participant movement tolerances seen with recent mobile EEG systems, allowing for real-time capture of neuronal activity (Freeman & Quiroga, 2013; Sipp et al., 2013). However, due to the mediums in which signals must travel through (skin and hair of the scalp, and cerebrospinal fluid of the brain) and back again, it suffers greatly from low spatial resolution and image quality of its imaging counter parts (Freeman & Quiroga, 2013). Finally, while filters and proper set-up of EEG may allow for clearer activity, artifact removal from environmental and biological factors (head and eye movement), can be time consuming (Freeman & Quiroga, 2013)

Post-synaptic potentials represent the origin of EEG signals; specifically, Excitatory Post-synaptic Potentials (EPSPs) (Holmes & Khazipov, 2007). EPSPs can propagate through the mediation of the excitatory neurotransmitter glutamate. When an action potential reaches the axon terminal the neuron releases glutamate, with the binding of glutamate to the post-synaptic receptor, sodium (Na^+) channels to enter the cell, thereby depolarizing the cell (the cell begins to trend to a more positive state) (Holmes & Khazipov, 2007). Each cell may receive multiple EPSP's either temporally (continuous from a parallel cell), or spatially, from neighboring pre-synaptic neurons, these potentials will begin to compound and if a critical threshold of EPSPs is reached then an action potential is generated (Holmes & Khazipov, 2007). Afferent inputs generated by EPSPs generate a depolarization within the cell body; resulting in a short timescale dipoles of a negative energy potential (μV) are then detected by a scalp electrode (Holmes & Khazipov, 2007).

EEG activity is obtained via scalp electrodes which detect diminished extracellular current flow from pyramidal neurons situated perpendicular to the skull (along with other non-cortical matter) and a surface electrode. Due to the location of the scalp electrodes, only vertically oriented dipoles can be measured using scalp EEG, which suggests that the only activity detectable by surface EEG are located in the superficial cortical sheet underneath the skull. Yet while, this measured electrical activity is subject to pollution from sources outside the head (e.g., lighting fixtures and muscle activation); a dimeter of 3-5mm of cortex may contain up to 10^6 neurons and 10^{10} synapses (P. L. Nunez, 1995). Further, the rhythms obtained through EEG recordings represents synchrony of numerous underlying cortical neurons. While thalamocortical circuits are indispensable for successful motor control (Alexander, 1994; Purves et al., 2012), the majority of the activity in question is believed to represent activity of cortical neurons that are close cortical structures of the electrodes (e.g., frontal, somatosensory, and primary motor cortex) (Holmes & Khazipov, 2007).

The electrical activity measured by EEG can be decomposed into its constituent oscillatory activity commonly defined by their frequencies. Some frequency bands that are particularly important for underlying brain circuits responsible for balance are the Delta (≤4 Hz), Theta (5-8 Hz), Alpha (8-13 Hz), Beta (13-15 Hz), and Gamma (30-50 Hz) (Bradford et al., 2016; Clark et al., 2020; Gwin et al., 2010; Severens et al., 2012; Sipp et al., 2013). Of particular interest may be the slow oscillatory activity of the slow rhythms of Delta and Theta. In order to receive any signals, several components and protocols must be used. Firstly, is the application of electrodes to the scalp of the head. These electrodes (often already embedded into a cap to be placed on the head), are of a 5mm diameter and are composed of a sliver chloride conductive material to allow for a conductive medium from the scalp to the electrode. An additional conductive solution is placed in between the space of the scalp and electrodes (e.g., a gel or even saline solution). This solution also has the benefit of reducing the electrical resistance (i.e., impedance) of the produced signal (as impedance decreases current increases). Proper signal-to-noise ratio impedance should be no more than 50 kΩ. (Beeck & Nakatani, 2019). While the signal in question should determine the electrode of study, it is often difficult to determine this location from a surface level. Hence, the EEG and electrodes are often organized to cover the entire head with a specified system. The 10-20 system is the most common and consists of six divisions that compose of 10 and 20% of the circumference of the head (Beeck & Nakatani, 2019). Additionally, since the eye can function as an additional electrical source, and further confound the output signal, additional electrodes are placed around the eyes of a participant in order to measure this output and subtract it from the whole (Beeck & Nakatani, 2019). Furthermore, as the electrical activity that is measured the scalp is of incredibly low voltage. An EEG amplifier is used to increase the signal of the weak signals and can increase the frequency of these signals by .01 to 50 Hertz (Hz) (Beeck & Nakatani, 2019). Additionally, an amplifier ground electrode must be employed as to provide a reference that can be used for rejection of confounding biological and environmental electric and magnetic fields (M. D. Nunez et al., 2016; P. L. Nunez & Srinivasan, 2006). Finally, the sampling rate of each of the specified frequencies should be considered. Considering that these signals are represented as a sine wave (two components, peak and trough), too low of a sampling rate while report incomplete data and thusly not represent the original signal (Beeck & Nakatani, 2019). A common was in which to acquire a proper sampling rate is to use the Nyquist theorem, which states that the sampling frequency should be twice the frequency of the signal. Therefore, since EEG frequencies range from 0.1 to 80 Hz, the sampling rate should be at least 160 Hz (Beeck & Nakatani, 2019).

Electroencephalography Protocols

The electrical potentials that are recorded through EEG proceed through brain matter, cerebrospinal fluid, and skull and scalp matter of the head before they are recorded at the scalp. Thusly the output signal, results in a complex electrical field and reflects electrical activity of many different sources (Beeck & Nakatani, 2019). While EEG data does seem appropriate for investigation into neural contribution in movement, the largest barrier may be an increased amount of background noise that may wash-out cortical activity (Beniczky & Schomer, 2020). Nonetheless, with proper filtering a processing steps, EEG has been shown to be effective in identifying cortical sources of activity (Bulea et al., 2015). A common and often first step in "cleaning" EEG data is by using basic frequency filters (Beniczky & Schomer, 2020). Two of these common filtering techniques are high-pass and anti-aliasing filters. A high-pass filter is intended to remove pre-specified low voltage changes. The low voltage changes are often correlated with skin potentials and EEG line noise (Beniczky & Schomer, 2020). Additionally, other confounding frequencies are produced by muscle activity in the scalp of the head, these frequencies may be eliminated using anti-aliasing filters (Beniczky & Schomer, 2020). However, besides environmental noise, the inherently poor spatial resolution in EEG makes source interpretation difficult (Freeman & Quiroga, 2013). Further, potentials that may be generated from non-cortical sources (e.g., muscular activation), may confound cortical source analysis (M. D. Nunez et al., 2016). However, a the relatively new method of Independent Component Analysis (ICA), may allow for separation of mixed signal output into independent clusters (Gwin et al., 2010). ICA can be used by applying an unmixing matrix to linearly mixed source (raw EEG data), allowing for recovery of the original source. Essentially, ICA is used to filter non-brain components from a signal. ICA has been previously assessed by Gwin et al., (2010) and was found successful for removing channel-based artifacts. For instance, channels are rejected (or marked for removal as they likely contain non-brain components), dependent upon certain preset parameters. These include a correlation threshold in which a channel is rejected if its correlation to other spatially near channels fails to meet the preconfigured value. In 2010 Gwin and colleagues demonstrated the process of ICA using a two-step approach. Essentially, time locked mechanical perturbances in the signal were identified (using regression analysis) and then applied to an unmixing signal, which was used to subtract out any similar recorded artifacts (Gwin et al., 2010). Additionally, ICA may also assist in cleaning the EEG signal from movement artifacts and present good benefit to spectral analysis (Gwin et al., 2010). (Beniczky & Schomer, 2020; Blinowska & Durka, 2006; Freeman & Quiroga, 2013).

The electrical activity measured by EEG can be represented by specific frequency bands Delta (1-4 Hz), Theta (4-7 Hz), Alpha (8-12 Hz), Beta (12-30 Hz), and Low Gamma (30-50 Hz). The spectral power (PSD) contained in these frequency bands is believed to represent distinct neural interactions and types of information processing from different cortical layers and structures (Blinowska & Durka, 2006), collectively representing dynamic energy states of the brain (Garcia et al., 2020). Faster rhythms are associated with more focal information processing (e.g., spiking), whereas slow wave oscillatory activity (delta/theta) are believed to arise from synaptic interactions (Sanchez-Vives, 2020), orchestrating spatially and temporally long-range cortical interactions, and plausibly represent motor planning and error detection (Chu et al., 2012; Combrisson et al., 2017; Lakatos et al., 2008; Sanchez-Vives, 2020; Sanchez-Vives et al., 2017). Both delta and theta PSD has been associated with postural control dynamics and performance (Aubonnet et al., 2022; Hülsdünker et al., 2015; Oliveira et al., 2017; Sipp et al., 2013; Slobounov et al., 2009). For instance a shift in attention during a task by suppressing unimportant neural activity during said tasks, has been previously associated with changes in delta PSD, (Harmony et al., 1996; Ibitoye et al., 2021); with an increase in delta activity being related to an increase in balance task difficulty (a more unstable posture), (Aubonnet et al., 2022; Ibitoye et al., 2021; Ozdemir et al., 2018). Additionally, theta frequency PSD has been previously found to change in a similar manner to the delta rhythm displaying an increase in PSD during the maintenance of balance under challenging conditions, and may be associated with sensorimotor processes (Barollo et al., 2020; Edwards et al., 2018; Hülsdünker et al., 2015; Sipp et al., 2013; Slobounov et al., 2009).

Electroencephalography and Neuromotor Control

Neuromotor control reflects integration of sensory and motor information in the central nervous system (CNS). Balance (a neuromotor task) is the ability to maintain an upright posture of one's body. Balance is accomplished by maintaining one's center of mass within theirs base of support. Several variables operationalized from standardized balance test have been shown to be useful for monitoring changes in neuromotor performance due to disease (Goble et al., 2014) and head trauma (e.g., Concussion) (Guskiewicz et al., 1996). Although the utility of these metrics stem from high reliability, measuring balance alone does not permit an examination of the neural processes required to perform the task. EEG allows for the examination of brain activity during a neuromotor task. EEG is a measurement of electrical activity on the scalp, which is primarily generated by neuronal activity from the brain (M. D. Nunez et al., 2016).

While an abundance of previous work has shown that balance tasks and EEG has allowed for the examination of neuromotor control, independently, the combination of these two methods has recently been used. Further, new EEG devices allow for a greater degree of body movement, which can provide enhanced temporal resolution of possible neural correlates associated with human movement and balance. In 2009, S. M. Slobounov and colleagues, examined the neural processes in participants during a loss of balance task. Participants were instructed to stand on one leg for as long as possible until failure of balance (i.e., repositing the supporting foot), with the balance measure was operationalized as virtual time to contact. Virtual time to contact is considered a balance metric that considers kinematic properties (i.e., velocity, displacement, and acceleration) as the center of pressure (CoP) of the supporting leg reaches its boundary of support (Slobounov et al., 1997, 1998). Additionally, EEG theta band activity was recorded and time-locked with the virtual time to contact metric. The authors reported not only an increase in virtual time to contact measures at the transition point to balance failure, but also a significant increase in theta PSD during this transition, specifically at the central region of the scalp. Further using maps of current sources, the authors suggested that these scalp-measured potentials may have originated in the anterior cingulate cortex (ACC) (Slobounov et al., 2009). This is consistent with previous research showing an interaction between the pre-frontal cortex and the ACC in monitoring error and guiding compensatory behavior (Botvinick et al., 2001; Gehring & Knight, 2000).

The results of S. M. Slobounov and colleagues have also been replicated in other research as well (Hülsdünker et al., 2015; Sipp et al., 2013). In 2013 Sipp and Colleagues, investigated neural correlates during the loos of balance while participants walked along a balance beam (low stability) compared to walking on a treadmill (high stability). For this study 24 participants (12 females, and 12 males, ± 5 yrs.), walked on a 2.5 cm wide ad 2.5 cm tall balance beam, with a loss of balance (stepping off of the balance beam) measured through motion capture; EEG measures were collected by a 256-elctrode EEG cap. Sipp et al, (2013), found an increase in theta PSD in the fronto and fronto-central regions of the scalp, indicating possible involvement of the ACC (Sipp et al., 2013) during error detection (e.g., loss of balance). Thus, suggesting the involvement of additional cortical components working in concert with the primary motor cortex (Sipp et al., 2013). Additionally, the work of Oliveira and colleagues (2017) reported increases in theta PSD when participants walked on a treadmill with their eyes closed compared to walking with their eyes open, at the fronto-central areas of the scalp, with the authors suggesting involvement of the pre-frontal and premotor cortex (Oliveira et al., 2017). These results indicate that a lack of visual input is associated with an increase in Theta PSD, (Oliveira et al., 2017). While a considerable amount of work has shown changes in theta band activity associated with balance tasks, the delta frequency band activity (1-4 Hz) may also be relevant when investigating the response to unstable balance. A study by Aubonnet and colleagues (2022) found an increase in delta PSD across conditions in participants who were asked to balance on a movable platform while a visual stimulus representing being on a boat at sea was presented, with delta PSD increasing along with balance task difficulty. Of particular importance is, delta PSD increased over the prefrontal brain areas, which have been implicated in other studies using source localization techniques, suggesting an increase in increased involvement of the frontal areas during unstable balance (Aubonnet et al., 2022)

Additional work by Hülsdünker et al., (2015), may support the idea of ACC and the fronto to fronto-central areas of the brain involvement during corrective balance control. In this study, 37 participants were asked to remain upright upon an oscillating platform across nine conditions of increasing difficulty. These conditions included: bipedal to unipedal support (dominant and nondominant legs) and a progression from solid to increasing unstable platform support; with fewer platform oscillations indicating better balance performance. Similar to previous studies, the results of this study displayed an increase in theta power at the fronto-central regions of the cortex with an increase in balance task difficulty (Hülsdünker et al., 2015). Moreover, significant positive correlations ($r144 = 0.31$, $p < 0.001$) were observed between platform oscillations and cortical theta activity of the fronto-central region of the scalp. In other words, as platform oscillations increased (indicating an increase in instability) so too did theta power, possibility indicating increased error signaling from an increase in balance stability (Adkin et al., 2006; Hülsdünker et al., 2015).

The work of Hülsdünker et al., (2015), combined the process of continual instability along with the investigation of cortical activity changes and adaptation. However, their work did not directly employ the use of sensory perturbation. The work of Barollo and colleagues, (2020), sought to examine this issue through the use of vibratory stimulation of the posterior lower leg muscles (to elicit a proprioceptive perturbation (Eklund, 1972), during a bipedal stationary balance task. Balance performance was operationalized as sway path length in the anterior-posterior, and lateral directions, with increased sway path length indicating poorer balance performance. Further cortical activity was recorded simultaneously using a high density (256-channel) EEG device, under two balance task conditions, eyes open and eyes closed. The results of this study revealed an increase in theta spectral power over the frontal-central region of the brain in both the eyes open and eyes closed conditions during the proprioceptive perturbation when compared to no perturbation (quiet stance), however, the eyes open condition did display greater spectral power across more regions of interest during the eyes open condition. Interestingly the increases in the theta frequency power may suggest not only error detection (Adkin et al., 2006), but also the am increase in processing of balance state (Slobounov et al., 2009). Additionally, the increase in theta band power may also suggest the planning of balance correction strategies or the analysis of an unstable state (Barollo et al., 2020). However, while correlation analysis between kinematic measures of balance performance (sway path length) and cortical activity was attempted, no significant correlations were found, with the authors suggesting that perhaps different balance measures may be indicated (Barollo et al., 2020). Nonetheless, this work may support that with increased demand of a balance task or perturbation of sensory systems, may lead to increased cortical activity that can be measured within the frequency of slow cortical oscillations (Barollo et al., 2020).

The current state of EEG collection for the examination of neuromotor control is an emerging concept. While current research is promising, there are considerations that should be noted. A common limitation seen in the literature reviewed was the criticism that the motor task in question may not have had the requisite level of difficulty to detect more reliable detection of input from supraspinal systems (Bradford et al., 2016; Gwin et al., 2011; Sipp et al., 2013). Further while the aforementioned work has provided insight into neural correlates during movement, the efforts in this space have largely focused on static balance tasks. Thus, a gap in the literature is the manner in which neural correlates change (or not) during a dynamic balance task. Given that “real world” tasks are typically dynamic in nature, adopting a dynamic balance test provides greater ecological validity to evaluate neuromotor function (Basford et al., 2003; Chou et al., 2004; Guskiewicz, 2001; Kuznetsov et al., 2018; Rhea et al., 2017). However, the current research did display consistency in results pertaining to the likely cortical components of neuromotor function and adaptation as well as, the frequency component of interest. Suggesting that these assessments may be useful for the investigation of further study (Clark et al., 2020).

CHAPTER III: OUTLINE OF PROCEDURES

Participants

Twenty-four participants enrolled in this study. Participant demographics are presented in Table 1. Participants were recruited through local colleges and the greater community of Greensboro, North Carolina. All participants were excluded if they failed to meet any of the following inclusion criteria: no history of neuromuscular condition, no history of a brain injury resulting in loss of consciousness, no history of a concussion 12 months prior to study participation, no family history of epilepsy, and no current use of psychoactive recreational or prescription drugs. This study was approved by the University of North Carolina at Greensboro's Institutional Review Board.

Table 1. Participant Demographics

Sex	n	Height (cm)	Weight (kg)	Age (yrs)
Female	17	163.6	62.2	22.5
Male	7	173.9	79.5	26.0
Mean		166.6	67.2	23.5
SD		9.3	13.6	4.6

Instrumentation

Each Participant performed the six balance tasks of the AccWalker TARGET protocol. AccWalker is a novel Android-based application that allows for the measurement of the movement of a participant's movement and orientation around three axes: pitch, yaw, and roll (recorded at 96 Hz). The app uses a standard Android sensor fusion algorithm called rotation vector to combine data from the phone's accelerometer, gyroscope, and the geomagnetic field sensor to calculate orientation. These data allow for the capture of the orientation of the thigh segment during stepping-in-place and then derive a series of movement parameters such as mean stride time, stride time variability, range of motion of the thigh, and variability of the thigh movement. The full protocol calls for three conditions for stepping in place: with eyes open (EO), with eyes closed (EC), and while moving head left-right, in a "Shaking No" manner (HS). The AccWalker was placed on the lateral thigh of the dominant thigh (participants were asked which leg they lead with when climbing stairs) midway between the center of the hip and knee joint. The entire AccWalker protocol (TARGET) consisted of two practice sessions (each 30 s) of the EO and HS conditions, two trials each (all 70 s) of the EO, EC, and HS conditions. In all conditions, participants stepped in place to a 10 s metronome embedded in the app, after which they were instructed to continue stepping at that pace for the final 60 s of the trial.

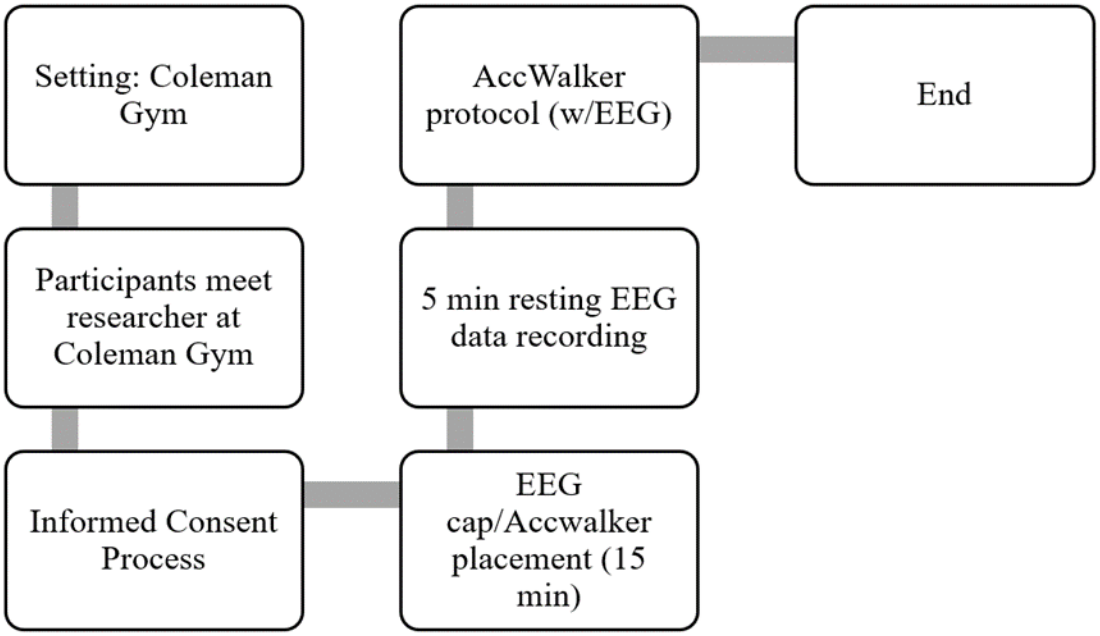
EEG data were collected using a 32 electrode R-Net EEG cap (Brain Products, Zeppelinstraße, Germany), sampled at 500Hz, and recorded in Brain Vision Recorder (Brain Vision, Morrisville, NC, USA). Participants sat quietly and as still has possible for 5 minutes before the start of the TARGET protocol. We continuously recorded EEG data at the start of rest and through the entire Target protocol (approximately 15 minutes total). The beginning and end of each TARGET trial was logged by a "1" (Begin) or "2" (End) by a keyboard strike on the computer. Time measurement for the beginning and end of each trial were also logged manually in the event of the keyboard strike being logged incorrectly or omitted.

The soccer kicking activity used a standardized soccer ball (400 g; 70 cm circumference; inflated to 12 psi) projected at a velocity of 38.7 ± 2.1 kph from a launcher (JUGS Sports International, Tualatin, USA) towards the participants positioned six meters away to simulate average soccer gameplay settings (Broglio et al., 2004; Haran et al., 2013). Participants were instructed to "do your best and kick the soccer ball back towards the launcher," in which the ball is redirected parallel to the initial path. The task will consist of 10 consecutive kicks over a ten-minute period.

Experimental Design

A flowchart of the experimental design is shown in figure 1 (below). All data collection was completed within the Coleman research gym at the University of North Carolina at Greensboro's (UNCG) main campus. Participants met with the researcher at the Balance and Training Laboratory, where the experiment was formally explained, and an informed consent form was signed. The participants were then brough to the research gym. Then the EEG cap and instrument harness were placed on the participant, as well as the AccWalker placed on the lateral thigh of the dominant leg. After each system was checked to be properly fitted and to be functioning, five minutes of resting EEG data was recorded. After this initial recording, participants performed the AccWalker protocol (pre-test). The entire AccWalker protocol consisted of two practice sessions (each 30 s) of the EO, and EC conditions, two 70 s EO conditions, two 70 s EC conditions, and two 70 s HS conditions (each of these conditions requires participants to step-in-place to the beat of an initial timed tone, which ceases after 5 s). After the initial testing session, the AccWalker and EEG amplifier (the EEG cap remained on the participant's head) was removed from the participant and the participant then performed the soccer ball kicking protocol. After the soccer kicking task, the AccWalker and EEG amplifier were then re-applied to the participant, five minutes of resting EEG data was recorded, and then the AccWalker protocol was then performed again (post-test).

Figure 1. Experimental Flowchart of Procedures



EEG Preprocessing

EEG data was recorded during each AccWalker session using a 32-Channel active electrode array (down sampled to match the AccWalker sampling rate of 96 Hz). During data collection the EEG amplifier was secured in a backpack to minimize motion artifact. Briefly, all data analysis was performed using MATLAB (MathWorks, Natick, MA), using the EEGLAB plugin (open-source environment for the processing of EEG data) using custom Matlab scripts. Independent component analysis via an ICA (Independent Component Analysis) was applied to pre-cleaned channel time series and again after cleaning and artifact removal.

AccWalker Preprocessing

The AccWalker data was pre-processed using custom Matlab scripts to extract the aforementioned temporal and spatial variables for each of the three conditions (EO, EC, HS). Per the TARGET protocol, two trials were performed for each condition, which were averaged for reliability purposes. The data were then exported to SPSS (v. 25, IBM, Armonk, NY).

Statistical Analysis

Purpose #1: To investigate the reliability of the AccWalker app as a test for neuromotor performance before and after light athletic activity (kicking a soccer ball).

Primary hypothesis #1 Temporal and spatial characteristics of dynamic balance will not significantly change between pre- and post-soccer kicking activity.

To examine the reliability of the AccWalker variables pre- and post-soccer kicking two analyses were performed, a Bland-Altman Reliability analysis (Bland & Altman, 2010; Giavarina, 2015) and the intraclass correlation (ICC(2,k)) and Standard Error of the Mean (SEM)

Purpose #2: To compare EEG spectral power characteristics of dynamic balance across three different AccWalker conditions.

Primary hypothesis #2 EEG power spectral density within the delta and theta frequency bands will increase across the three AccWalker conditions.

Delta and Theta PSD were calculated as a percentage of all other band activity (Broadband, BBDelta and BBTheta) (e.g., $BBPSD = \frac{PSD\ Delta}{Delta+Theta+Alpha+Beta+Gamma}$. SPSS version (v. 25, IBM, Armonk, NY) was used for the final analysis of the processed Delta and Theta PSD. Separate, repeated measures ANOVAs (RM-ANOVAs) tested the effect of condition (4 conditions: Rest, Eyes Open (EO), Eyes Closed (EC), and Head Shake (HS)) on PSD in each band (delta, theta). Degrees of freedom were corrected (Hunyh-Feldt) in cases where the assumption of sphericity was violated. Effect sizes are reported as partial-eta squared (ηp2). Significant main effects were decomposed by pairwise comparisons, and Type I error rate was controlled by Bonferroni correction for six comparisons $\left(\frac{p-value}{6}\right)$.

Purpose #3: To examine correlations between EEG spectral power characteristics and temporal and spatial kinematic data during a stepping in place task (AccWalker TARGET protocol).

Primary hypothesis #3 EEG PSD within the delta and theta frequency bands will correlate with the temporal and spatial kinematic variables measured using the AccWalker TARGET protocol.

Before analyzing, the individual values for BBDelta, BBTheta, and the temporal and spatial variables measured from the AccWalker device were z-score transformation to account for individual differences $z = \frac{(x-\mu)}{\sigma}$ within Microsoft Excel and a Shapiro-Wilk test of normality was performed using SPSS (v. 25, IBM, Armonk, NY). The test of normality indicated that our data may not be normally distributed (see. Appendix C), so we elected to perform Spearman nonparametric correlations for our correlation analysis. Spearman's rho was calculated between the temporal and spatial variables collected via the AccWalker and BBDelta and BBTheta for each condition.

CHAPTER IV: MANUSCRIPT I

Introduction

A concussion is a type of traumatic brain injury which can occur when a significant external force is directed at the head, leading to an abrupt change in velocity of the head resulting in linear and/or rotational movement of the brain within the skull (Centers for Disease Control and Prevention (CDC), 2019). In 2013, approximately 2.8 million concussion related emergency department visits occurred in the U.S. (Taylor et al., 2017). Considerable research has investigated the acute and long term-effects of concussions on brain health. However, there is relatively less research on the identification of injury from repetitive sub-concussive impacts. A sub-concussive impact is a head impact that does not produce the clinical symptoms associated with a concussion (Belanger et al., 2016; Caccese, 2018; Mainwaring et al., 2018). While the sub-concussive impacts may not be considered as injurious as that of concussions (Belanger et al., 2016; Mainwaring et al., 2018), repetitive exposure to sub-concussive impacts has been shown to be associated with both acute (Caccese et al., 2018; Di Virgilio et al., 2016; Haran et al., 2013; Hwang et al., 2017; Kaminski et al., 2007; Rhea et al., 2017) and chronic (Bailes et al., 2013; Downs & Abwender, 2002; Manning et al., 2020) negative outcomes in neuromotor and neurocognitive ability.

Neuromotor control reflects the integration the sensory systems of the central nervous system (CNS) and peripheral neuromuscular system. Balance (a neuromotor task) is the ability to maintain an upright posture of one’s body. Balance is accomplished by maintaining one’s center of mass within one’s base of support. However, whether in a static or dynamic state, the upright posture is innately unstable, as any small shift of the center of mass’ position within the base of support can result in instability and thusly a fall (unless additional body maneuvers are quickly incorporated). Hence, constant monitoring of the position of the body within space is needed by three main sensory systems; the somatosensory (proprioceptive), visual, and vestibular systems in a fluctuating manner (Peterka, 2018). The proprioceptive, visual, and vestibular systems are organized throughout the body and can work in concert with each other to meet five tasks: (1) resist gravity to maintain and upright posture, (2) maintain the body's center of mass within its base of support, (3) provide for postural stability of the task, (4) provide for acceptable foot-ground clearance during dynamic activities, and (5) control for the accelerations of the head to stabilize visual and vestibular systems (Winter, 1989).

Balance assessment (postural control) is considered a best practice in the management and detection of concussion, repetitive head trauma, and other neurological conditions (Guskiewicz et al., 1996; Riemann & Guskiewicz, 2000). There are a variety of balance assessments that can be employed to assess neuromotor performance and these assessments can be separated by either static or dynamic balance assessments, or objective or subjective assessments. The choice of an assessment should be guided by the demands resources available as dictated by the environmental setting and research question.

A static balance test, until recently, required the use of high-cost force plates in a specially designed laboratory space (Schmidt et al., 2012). However, recent technologies have come to market that provide a low-cost alternative and portability (Goble et al., 2016; Patterson et al., 2014). Static balance tests are classified as such because, during each assessment the participant is required to remain as still as possible in the testing position. To operationally define static balance, either CoP displacement or other forms of displacement are determined from the initial starting point (position at start) and are continually recorded throughout the test. Contrary to static balance assessments, dynamic balance assessments may allow for increased resolution of alterations of the three sensory systems (Basford et al., 2003; Chou et al., 2004; Goble et al., 2016; Guskiewicz et al., 2001; Kuznetsov et al., 2018; Rhea et al., 2017). Through the use of a standard movement protocol (e.g., stepping-in-place) over several conditions (eyes open, eyes closed, or complimentary head movement whilst stepping-in-place), not only can smaller deficits in balance be assessed, but alterations in the three sensory systems may be elicited (Kuznetsov et al., 2018; Rhea et al., 2017). Common variable data from dynamic tests include RoM and rhythm consistency (or variability of that rhythm), and the differences in this data pre- and post- either a neural injury or therapeutic modality may suggest alterations in the three sensory systems (Rhea et al., 2017).

The choice between the use of static, dynamic, subjective, and objective balance assessments should be determined by the needs of the research question or reason for assessment (Rhea, at al., 2018). Further, while subjective tests may allow for a more economical assessment, they may be subject to lower sensitivity and reliability (Murray et al., 2014). Additionally, while the current market does offer objective static balance tests, the relative simplicity of the task (requiring less vestibular input) may limit the ability to detect small changes in neuromotor control, whereas dynamic tests may be more sensitive in this context (Baloh et al., 1994; Rhea et al., 2017). If one requires the assessment of the body's ability to adapt to more "real-life" situations and perturbations of the three sensory systems, an objective and dynamic assessment should be selected (Rhea, et. al., 2018).

The AccWalker smartphone app can be used for this purpose. AccWalker is an Android based smartphone application and uses the accelerometers and gyroscopes of the smartphone to record the movement patterns of the dominant lower extremity during a stepping in place task (a dynamic balance activity, (Kuznetsov et al., 2018)). This data is used to calculate several spatial and temporal characteristics of gait timing as a measure of neuromotor control. Previous use of the AccWalker device as shown acceptable reliability and validity compared to motion capture of treadmill walking (Kuznetsov et al., 2018), resistance to fatiguing exercise (Stafford et al., 2020), and identification of decreased neuromotor performance from blast exposure from military setting explosions (shoulder mounted weaponry) (Rhea et al., 2017). These qualities may allow the AccWalker to be useful for the detection of affected neuromotor performance by sub-concussive repetitive impacts in the sport setting (e.g., soccer heading). This is important, as current literature has shown mixed results in neuromotor changes after a repetitive soccer ball heading task (Broglio et al., 2004; Caccese et al., 2018; Di Virgilio et al., 2016; Haran et al., 2013; Hwang et al., 2017; Kaminski et al., 2020; Schmitt et al., 2004). These different results could be due to sensitivity differences in the neuromotor tests used. However, before adoption AccWalker in this context, it is important to first demonstrate its reliability in such an environment. The purpose of this study was to investigate the reliability of the AccWalker app as a test for neuromotor performance before and after light athletic activity (kicking a soccer ball). Our hypothesis was that the temporal and spatial characteristics of dynamic balance will not significantly change between pre- and post-soccer kicking activity, thus providing evidence of its reliability in this context.

Methods

Participants

Twenty-four participants enrolled in this study. Participant demographics are presented in Table 2. Each participant was assessed pre- and post-a soccer kicking task. For this task participants were asked to return each ball by kicking the ball back to its origin. Kicking a soccer ball was chosen to allow for the identification of cortical activity changes brought about from exercise (Brümmer et al., 2011). Participants were recruited through local colleges and the greater community of Greensboro, North Carolina. All participants were excluded if they failed to meet any of the following inclusion criteria: no history of neuromuscular condition, no history of a brain injury resulting in loss of consciousness, no history of a concussion 12 months prior to study participation, no family history of epilepsy, and no current use of psychoactive recreational or prescription drugs. This study was approved by the University of North Carolina at Greensboro's Institutional Review Board.

Table 2 Participant Demographics

Sex	n	Height (cm)	Weight (kg)	Age (yrs)
Female	17	163.6	62.2	22.5
Male	7	173.9	79.5	26.0
Mean		166.6	67.2	23.5
SD		9.3	13.6	4.6

Experimental Design

This study used a pre-test/post-test design. Both pre- and post-testing included using the mTBI Assessment of Readiness Gait Evaluation Test (TARGET) AccWalker protocol (Rhea et al., 2022; see Experimental Procedure section for details) before and after kicking ten soccer balls. The kinematic variables of interest consisted of both spatial and temporal measures. The temporal measures were Mean Stride Time (measured in seconds) and the coefficient of variation (CV) of stride time (measured as a percentage). Stride time in general was measured as the duration between maximal thigh flexion peaks in the angular displacement time series. Mean Stride Time was calculated by averaging all stride times that were sampled (at 96 Hz) per trial. Stride Time CV was calculated by as the quotient of the standard deviation of sampled stride times to the Mean Stride Time, and then converting this value to a percentage (multiplying by 100). The spatial measures included the CV of Peak Thigh RoM, Thigh RoM, and the standard deviation (SD) of the Peak Lift and Return Velocities. Peak Thigh RoM CV was calculated similar to Stride Time CV but using the measured the Thigh RoM values standard deviation to its respective mean. Thigh RoM was gathered by the smartphone itself via the native measurement devices (accelerometers, magnetometers, and gyroscopes), which then placed into a fusion algorithm, creating a mean value of all measured relative angle displacements of the phone at the measured thigh's peak flexion at the hip. Further, Peak Lift Velocity SD and Peak Return Velocity SD were obtained again via the smartphone as represented the standard deviation of the peak velocity during leg lift and return velocity. These measures were previously defined and showed good reliability and correlation to treadmill walking (Kuznetsov et al., 2018).

Instrumentation

AccWalker is a novel Android-based application that allows for the measurement of the movement of a participant's movement and orientation around three axes: pitch, yaw, and roll (recorded at 96 Hz). The app uses a standard Android sensor fusion algorithm called rotation vector to combine data from the phone's accelerometer, gyroscope, and the geomagnetic field sensor to calculate orientation. These data allow for the capture of the orientation of the thigh segment during stepping-in-place and then derive a series of movement parameters such as mean stride time, stride time variability, range of motion of the thigh, and variability of the thigh movement. The full protocol calls for three conditions for stepping in place: with eyes open (EO), with eyes closed (EC), and while moving head left-right, in a "Shaking No" manner (HS). The AccWalker was placed on the lateral thigh of the dominant thigh (participants was asked which leg they lead with when climbing stairs) midway between the center of the hip and knee joint. The entire AccWalker protocol (TARGET) consisted of two practice sessions (each 30 s) of the EO, and HS conditions, two 70 s EO conditions, two 70 s EC conditions, and two 70 s HS conditions (each of these conditions requires participants to step-in-place to the beat of an initial timed tone, which ceases after 5 s). The three conditions of the AccWalker protocol are each employed to measure neuromotor performance with altered sensory information from the three neuromotor systems (visual, proprioceptive, and vestibular). The AccWalker protocol has previously shown validity in assessing neuromotor behavior post sub-concussive exposure and is resistant to perceived fatigue from repetitive testing (Rhea et al., 2017; Stafford et al., 2020).

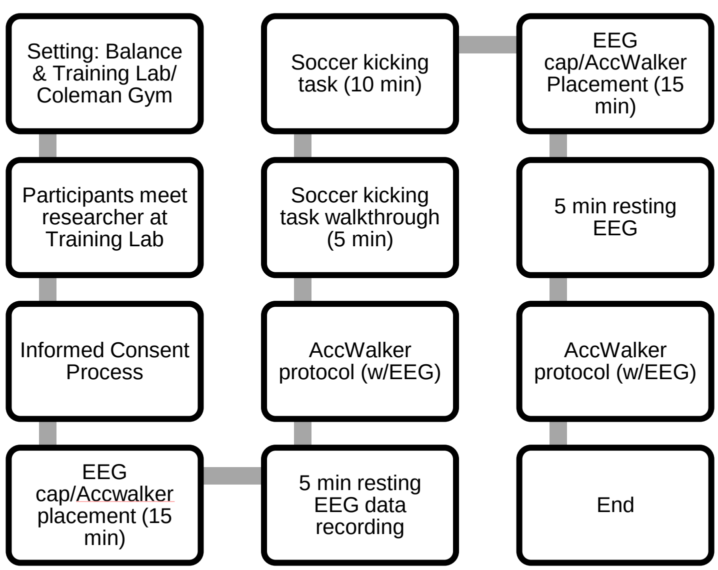
For the soccer-kicking activity, ten soccer balls (size 5, 70 cm circumference, 400g, inflated to 32 psi) were projected at a path perpendicular to the participant from six meters away (Broglio et al., 2004; Caccese et al., 2018; Di Virgilio et al., 2016; Haran et al., 2013; Hwang et al., 2017; Kaminski et al., 2020; Schmitt et al., 2004). Each soccer ball was projected at a speed of 10.75 meters per second; this velocity was determined by a literature review of studies using soccer-heading as a model for subconcussive impacts (Broglio et al., 2004; Caccese et al., 2018; Di Virgilio et al., 2016; Haran et al., 2013; Hwang et al., 2017; Kaminski et al., 2020; Schmitt et al., 2004). A Jugs (Jugs Sports, Tualatin, OR) soccer ball launcher was used with an angle set to parallel to the ground. Between each AccWalker testing sessions, ten soccer balls were projected towards the participants' feet at a rate of one ball/min over ten minutes; the participants were instructed to "do your best" at kicking the ball forward, back to the Jugs machine.

After the initial testing session, the AccWalker was removed from the participant and then the participant performed the soccer ball kicking protocol. After the soccer kicking task, the AccWalker was then refitted to the participant, and then the AccWalker protocol was then performed again (post-test).

Experimental Procedure

Participants met with the researcher in the research gymnasium, where the experiment was formally explained, and an informed consent form was signed. After a five-minute rest period, the AccWalker phone (Google Pixel) was affixed to the lateral side of the participants dominant thigh and the steeping in place movement was demonstrated for the EO condition and a 30s practice of this condition was performed. After this, the participant was asked to perform a 30s practice of the HS condition after a short demonstration. Then six full trials (2 x EO, 2 x EC, 2 x HS), lasting 70s each were performed, with the participant expressing that that were able to continue after each trial. After the initial AccWalker data collection, the participant then performed ten soccer ball kicks within ten minutes ($\frac{1\ ball}{min}$). After the soccer kicking session, the post AccWalker data collection was performed in the exact same way as the pre-soccer ball kicking condition; a flowchart of the procedure is in Figure 2.

Figure 2. Experimental Flowchart for Procedure



Statistical Analysis

The AccWalker data was pre-processed using custom Matlab scripts to extract the aforementioned temporal and spatial variables for each of the three conditions (EO, EC, HS). Then each trial (2) for each condition were averaged together and the means and standard deviations were calculated in Microsoft Excel. To examine the reliability of the AccWalker variables pre- and post-soccer kicking two analyses were performed, a Bland-Altman Reliability analysis (Bland & Altman, 1999, 2010; Giavarina, 2015) and the intraclass correlation (ICC(2,k)) and Standard Error of the Mean (SEM) analysis (Kuznetsov et al., 2018). These analyses were used to determine the not only the similarity of our measures to those of previous work (Kuznetsov et al., 2018), but also to provide for a comparison of the same kinematic data collected pre- and post-physical activity, allowing for increased validity of the AccWalker TARGET protocol in the use of neuromotor assessment before and head-impact exposure (Bland & Altman, 1999, 2010; Giavarina, 2015). Further, ICC(2,k) and SEM analysis were also used for three reasons; first for a comparison to prior study (Kuznetsov et al., 2018), second the ICC analysis to provide for a more robust analysis of the pre- and post- kinematic data as the Bland-Altman method was initially intended for the comparison of separate instrumentation, and third SEM analysis can allow for a measure of absolute reliability, which can be considered to be resistant to individual differences in performance (Weir, 2005). As our sample used no criteria based upon the neuromotor ability of the participants (with the exception of possessing the ability to step in place continuously for 10 minutes), the SEM measure allows for a true pre-post- comparison (Weir, 2005).

For the Bland-Altman analysis, differences between the pre and post were calculated in Excel (e.g., Pre - Post) as a well as the mean of the difference between Pre and Post conditions

($\frac{Pre-Post}{2}$). Prior to statistical analysis an outlier analysis was done (in MS Excel) in the following manner: data was trimmed in two ways. First participants were removed if their CV stride time was beyond three standard deviations (all data removed). Then participants were removed on a case-by-case basis (e.g., removed form Stride time CV, but not Thigh RoM) if the difference between the pre- and post- session were beyond three standard deviations from the sample. The cleaned data were evaluated using the Bland-Altman reliability method through SPSS version 28 (IBM, Armok, NY). To assess any potential bias between pre and post measures a one sample t-test was used to evaluate the differences in pre-and post- values ($p >$ 0.05, indicating no bias). Then a Bland Altman plot was constructed displaying the mean of the difference (x-axis) to the difference between pre- and post- conditions (y-axis). After plotting our data confidence intervals were set to 95% by multiplying the mean of the difference between pre- and post then adding and subtracting 1.96 multiplied by the standard deviation of the mean differences (Bland & Altman, 2010). Bias effect size was designated as the "Point Estimate" from the one sample t-Test, using the following criteria: 0.2 to 0.5 indicating a small effect size and equal to and greater than 0.5 indicating a large effect size (Howell, 2013).

ICC(2,k) and SEM were calculated using SPSSS version 28 (IRB, Armok, NY). A One-Way ANOVA was used for each pair (Pre and Post) of AccWalker variables and was used to calculate the ICC(2,k) (Kuznetsov et al., 2018). Additionally, SEM was calculated as the square of the error term from the ANOVA analysis for each pair of pre- and post- AccWalker variables (Weir, 2005). ICC(2,k) was selected as not only was this statistic previous shown to be a valid measure for this data (Kuznetsov et al., 2018), but both measures have been shown to account for both systemic and random error (Koo & Li, 2016; Weir, 2005). Additionally ,while prior research (Kuznetsov et al., 2018) has validated the use of these statistical measures previously, the use of these measures for the comparison in a pre- to post-test manner is unique. Our ICC(2,k) thresholds were selected based upon previous work and were as follows with > 0.75 indicating good test-retest reliability (Kuznetsov et al., 2018).

Results

Data from 24 participants was analyzed. AccWalker descriptive statistics are presented in Table 3, the Bland-Altman results, and ICC (2,k) and SEM results are presented in Tables 4 and 5, respectively. Our results showed that all spatial metrics (with the exception of peak lift velocity within the EO and EC conditions, and peak return velocity with the HS condition) displayed good test-retest reliability (ICC >0.75) and showed no estimate in the bias between pre- and post- conditions.

Table 3. AccWalker Descriptive Statistics

		AccWalker Descriptives Pre						AccWalker Descriptives Post					
		EO		EC		HS		EO		EC		HS	
	Unit	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Temporal Metrics													
Mean stride time	s	1.13	0.04	1.12	0.04	1.11	0.05	1.11	0.04	1.11	0.05	1.10	0.05
CV stride time	%	2.30	0.55	2.50	0.52	2.74	0.80	2.21	0.50	2.51	0.67	2.52	0.76
Spatial Metrics													
Peak thigh CV	deg	7.41	2.42	8.88	3.95	8.65	2.76	7.18	1.53	8.27	2.75	8.16	2.43
Thigh RoM	deg	38.41	10.26	35.56	10.78	33.12	10.08	34.85	9.71	31.53	9.16	30.16	8.08
Peak lift vel SD	deg • s^{-1}	13.81	3.77	14.45	3.10	14.25	3.15	12.95	3.09	12.94	3.21	13.02	2.30
Peak return vel SD	deg • s^{-1}	13.93	3.10	15.13	3.10	14.57	3.28	13.24	2.65	13.67	3.32	13.79	2.72

Table 4. AccWalker Bland Altman Results.

		Bland Altman																	
		EO						EC						HS					
	Unit	Bias	Bias *p-value*	Bias Effect Size	SD	Lower 95% CI	Upper 95% CI	Bias	Bias *p-value*	Bias Effect Size	SD	Lower 95% CI	Upper 95% CI	Bias	Bias *p-value*	Bias Effect Size	SD	Lower 95% CI	Upper 95% CI
<u>Temporal Metrics</u>																			
Mean Stride Time	s	0.02	0.03	0.47	0.05	0.002	0.045	0.01	0.13	0.32	0.04	-0.004	0.033	0.01	0.19	0.18	0.04	-0.010	0.025
Stride Time CV	%	0.08	0.20	0.27	0.31	-0.047	0.212	-0.01	0.94	-0.02	0.35	-0.152	0.141	0.23	<0.01	0.67	0.34	0.084	0.373
<u>Spatial Metrics</u>																			
Peak Thigh RoM CV	deg	0.23	0.43	0.16	1.39	-0.358	0.814	0.61	0.12	0.33	1.86	-0.176	1.391	0.49	0.11	0.25	1.94	-0.326	1.314
Thigh RoM	deg	3.55	0.01	0.58	6.13	0.966	6.142	4.04	<0.01	0.70	5.77	1.600	6.473	2.97	<0.01	0.59	5.01	0.850	5.083
Peak Lift Velocity SD	deg • s^{-1}	0.86	0.09	0.36	2.41	-0.155	1.877	1.52	<0.01	0.62	2.45	0.480	2.552	1.23	0.03	0.42	2.97	-0.022	2.486
Peak Return Velocity SD	deg • s^{-1}	0.69	0.25	0.24	2.85	-0.509	1.895	1.46	0.06	0.41	3.57	-0.048	2.967	0.78	0.12	0.24	3.20	-0.571	2.131

Table 5. AccWalker ICC(2,k) and SEM Values.

		AccWalker					
		ICC(2,k)			SEM		
	Unit	EO	EC	HS	EO	EC	HS
Temporal Metrics							
Mean Stride Time	s	0.41	0.73	0.80	0.03	0.03	0.03
Stride Time CV	%	0.81	0.91	0.84	0.22	0.24	0.24
Spatial Metrics							
Peak Thigh RoM CV	deg	0.87	0.92	0.84	0.98	1.31	1.37
Thigh RoM	deg	0.87	0.87	0.90	4.33	4.08	3.54
Peak Lift Velocity SD	deg • s^{-1}	0.85	0.78	0.56	1.70	1.73	2.10
Peak Return Velocity SD	deg • s^{-1}	0.68	0.34	0.79	2.01	2.52	2.26

Discussion

The purpose of this study was to investigate the reliability of the AccWalker app as a test for neuromotor performance before and after light athletic activity (kicking a soccer ball). Our hypothesis was that the temporal and spatial characteristics of dynamic balance will not significantly change between pre- and post-soccer kicking activity. The results of this study indicate that the AccWalker TARGET protocol displayed good test-retest reliability with similar data characteristics to previous work (Kuznetsov et al., 2018). The Bland Altman analysis revealed that the Peak Thigh RoM CV measure displayed no pre- post-test bias before or after soccer kicking. Furthermore, almost all metrics displayed greater than 0.75 ICC(2,k), with the exception of Mean Stride Time and PRVSD in the EO and EC condition. In addition, Peak Thigh RoM CV on average displayed an ICC(2,k) value of 0.87 indicating excellent reliability. This may be important, as previous work showed a decrease in Peak Thigh RoM CV in military personnel for up to 72 hours after explosive training in which they were subject to blast exposure equivalent to a sub-concussive force (Rhea et al., 2023). While blast exposure does present with different kinetic and kinematic characteristics than blunt force that may be seen in soccer heading or other sub-concussive impacts, gait disturbances have also been seen during recovery from blunt-force concussion in previous literature (Buckley et al., 2013; Catena et al., 2009; Oldham et al., 2016), which may represent the same characteristics as measured by the AccWalker.

Additionally, we noted thigh kinematic changes (notably Thigh RoM) across the different sensory conditions; Thigh RoM was reduced in both pre- and post-soccer kicking from the EO to the EC condition (-7.42˚ and -6,86˚, pre and post, respectively), and the EC to the HS condition (-9.53˚ and -4.35˚, pre and post, respectively). These characteristics align with observations when different sensory systems are perturbed during either gait or a balance task. Notably with visual system disrupted externally (e.g., dark room, or similar visual restriction), shorter and quicker stride frequency were found in participants (Hallemans et al., 2009; Oliveira et al., 2017). Additionally, through galvanic stimulation of the vestibular system, Hwang et al., (2015) found a decrease in leg angle during a walking stability task (Hwang et al., 2017).

Further, our kinematic data is consistent with previous reliability data (for previous data, see Kuznetsov et al., 2018). Specifically, our temporal metric data (Peak Thigh RoM CV, Thigh RoM, Peak Lift Velocity SD, Peak Return Velocity SD), displayed ICC(2,k) values compare well those of the previous work. This is valuable in a few ways; first, providing similar data to previous work using the same device displays that the AccWalker is a reliable measure of kinematic data, AccWalker is reliable for pre-post designs integrated with light physical activity, and compared to the previous work of Kuznetsov and colleagues in which the same data as this study was collected across days and was compared to motion capture metrics, this data suggest the same reliability for the measure of neuromotor performance during an acute time frame pre- and post-physical activity. These characteristics show that the that AccWalker can be used to evaluate the neuromotor performance of postural control.

However, this study is limited in one aspect. Specifically, we did not measure either mental or physical workload of the participant during the TARGET protocol. One such measure may have been, the use of the NASA Task Load Index (NASA-TLX) to measure the mental and physical workload associated with the TARGET protocol. However, previous work has integrated a fatiguing exercise and the NASA -TLX in a pre- post-design using the AccWalker device. The results of this study showed no correlation to NASA-TLX scores, and the kinematic characteristics measured by AccWalker (Stafford et al., 2020).

The results of this study indicate postural control assessment can be measured reliably in a pre- to post-test design. This may be important as there is a current lack of assessment for the effects of sub-concussive head impact exposure. While portable and relatively cost-effective devices are available, identification of a decrease in neuromotor performance may be lost due to the inherent limitations of these devices. For instance, Balance Tracking Systems (BTrackS), offer a portable force plate that can be interfaced to portable computer technology to reliable assess decrements to postural control from suspected concussive impacts to the head. However, as the neuromotor changes from sub-concussive exposure appear to present with increased nuance and transitory effects (Di Virgilio et al., 2016; Haran et al., 2013; Hwang et al., 2017) compared to concussive exposure, static balance testing may lack the appropriate resolution to detect sub-concussive changes in neuromotor performance and the body's ability to adapt to "real-life" (or more dynamic) situations (Baloh et al., 1994; Rhea et al., 2017, 2018).

CHAPTER V: MANUSCRIPT II

Introduction

Neuromotor control reflects integration of sensory and motor information in the central nervous system (CNS). Balance (a neuromotor task) is the ability to maintain an upright posture of one’s body and is accomplished by maintaining one’s center of mass within their base of support. Several variables operationalized from standardized balance tests have been shown to be useful for monitoring changes in neuromotor performance due to disease (Goble et al., 2014) and head trauma (e.g., Concussion) (Guskiewicz et al., 1996). Although the utility of these metrics stem from high reliability, measuring balance alone does not permit an examination of the neural processes required to perform the task.

Electroencephalography (EEG) allows for the examination of brain activity during a neuromotor task through the measurement of electrical activity on the scalp that, once cleaned, represents post-synaptic potentials from underlying brain circuits (M. D. Nunez et al., 2016) EEG-measured slow wave oscillatory activity (Delta, 1 – 4 Hz and Theta, 5 – 8 Hz) is often associated with postural control dynamics and performance (Aubonnet et al., 2022; Hülsdünker et al., 2015; Oliveira et al., 2017; Sipp et al., 2013; S. M. Slobounov et al., 2009). These rhythms are believed to arise from synaptic interactions (Sanchez-Vives, 2020), orchestrating spatially and temporally long-range cortical interactions, and plausibly represent motor planning and error detection (Chu et al., 2012; Combrisson et al., 2017; Lakatos et al., 2008; Sanchez-Vives, 2020; Sanchez-Vives et al., 2017). An increase in delta frequency activity has been associated with a transition from stable to increasing unstable balance (Aubonnet et al., 2022; Ibitoye et al., 2021; Ozdemir et al., 2018), which may be related to the suppression of unrelated neural activity (Harmony et al., 1996; Ibitoye et al., 2021). An increase in theta activity has been found under similar circumstances (Barollo et al., 2020; Edwards et al., 2018; Hülsdünker et al., 2015; Sipp et al., 2013; Slobounov et al., 2009), and may be associated with sensorimotor processes of the pre-frontal and premotor cortices (Oliveira et al., 2017). Although delta and theta activity may have similar underlying neural mechanisms, these two rhythms are not often combined into a slow cortical oscillations category. The separation of these two frequency bands may be because theta has been noted to represent a higher energetic state, compared to the delta frequency (Garcia et al., 2020). Further, the irregular nature of PSD may influence each band separately across different balance conditions, participants, or both (Donoghue et al., 2020; Finley et al., 2022).

These theoretical interpretations of EEG-measured slow rhythms are supported by a handful of studies that have paired EEG with balance tasks. One study demonstrated that theta power spectral density (PSD) over a central region of the scalp, approximating the somatomotor cortex, was increased during brief periods of instability. Source localization techniques further revealed that these changes may have originated in the anterior cingulate cortex (ACC) (Slobounov et al., 2009), which is responsible for monitoring error and guiding compensatory behavior through interactions with the prefrontal cortex (Botvinick et al., 2001; Gehring & Knight, 2000). These prefrontal-ACC interactions were revealed in a later study that reported increased theta PSD in the fronto and fronto-central regions of the scalp when participants traversed a balance beam compared to when they walked on a treadmill (Sipp et al., 2013).

Even relatively simple motor tasks can be made more challenging when sensory information is perturbed or impaired. It would be expected that such conditions would require even greater recruitment of cortical motor regions (i.e., prefrontal cortex, premotor cortex) working in concert with the primary motor cortex. Oliveira and colleagues (2017) reported enhanced fronto-central theta PSD when participants walked on a treadmill with their eyes closed compared to when they performed the same task with their eyes open. In a different study, participants who were asked to balance on a movable platform while a video was played to mimic the feeling of being on a boat at sea. Consistent with prior literature, delta PSD increased over the prefrontal brain areas with increasing task difficulty (Aubonnet et al., 2022).

While much of this previous work has provided good evidence of coordinated activity within and across distinct regions of the cortex, there is a dearth of work investigating neural correlates of slow wave oscillations during a continuous balance task. The use of continuous balance tasks is important, as this type of task better represents a more "real-world" representation of human balance (Rhea et al., 2018). Further, if the proposed pre-frontal-ACC interaction is associated with balance performance, then this association should be measurable. Hülsdünker and colleagues (2015), were able to confirm possible pre-frontal-ACC interactions during a continuous balance under nine conditions of increasing demand and instability. Moreover, significant positive correlations ($r144 = 0.31$, $p < 0.001$) were observed between platform oscillations and cortical theta activity of the fronto-central region of the scalp. In other words, as platform oscillations decreased so too did theta power, possibly indicating reduced error signaling from an increase in balance stability (Adkin et al., 2006; Hülsdünker et al., 2015). In addition, continuous perturbation of sensory systems may also allow for a better comparison to relevant balance tasks. This idea of proprioceptive integration at the cortical level during balance is supported by similar findings of increased fronto-central theta power during experimentally manipulated proprioception, an arguably stronger research design in which experimenters used a 256-channel High Density EEG device simultaneously perturbing proprioception by 85 Hz vibration of the posterior lower leg (Barollo et al., 2020).

While the aforementioned work has provided insight into neural correlates during movement, the efforts in this space have largely focused on static balance tasks. Thus, a gap in the literature is the manner in which neural correlates change (or not) during a dynamic balance task. Given that "real world" tasks are typically dynamic in nature, adopting a dynamic balance test provides greater ecological validity to test neuromotor function (Basford et al., 2003; Chou et al., 2004; Guskiewicz, 2001; Kuznetsov et al., 2018; Rhea et al., 2017). This acknowledgement was the impetus for the development of the AccWalker smartphone app and the creation of the mTBI Assessment of Readiness Gait Evaluation Test (TARGET) protocol (Rhea et al., 2022; see Experimental Procedure section for details). AccWalker is an Android based smartphone application and uses the accelerometer, magnetometer, and gyroscope of the smartphone to record the movement patterns of the dominant lower extremity during a stepping in place task (a dynamic balance activity (Kuznetsov et al., 2018). This data is used to calculate several spatial and temporal characteristics of dynamic balance control. Acceptable reliability and validity compared to motion capture of treadmill walking has been shown with AccWalker (Kuznetsov et al., 2018). Furthermore, AccWalker has been shown to be resistant to fatiguing exercise (Stafford et al., 2020) and has the ability to differentiate neuromotor performance after a blast exposure (Rhea et al., 2017). While AccWalker has shown clinical and research utility, neural correlates of the motion that the app is capturing remain unknown. This is of particular interest, as the TARGET protocol includes three conditions designed to probe sensory contributions during dynamic balance control. This consists of stepping-in-place with: (1) eyes open, (2) eyes closed, and (3) shaking the head from side-to-side. Understanding the EEG characteristics during the task—as well as the neural correlates of AccWalker performance with

EEG characteristics—would provide a deeper understanding on the dynamic neuromotor control process.

The purpose of study was two-fold: 1) to compare EEG PSD characteristics of dynamic balance across three different AccWalker conditions, and 2) to examine correlations between EEG PSD characteristics and temporal and spatial kinematic data during a stepping in place task (mTBI Assessment of Readiness Gait Evaluation Test (TARGET)). We had two hypotheses 1) EEG (PSD) within the delta and theta frequency bands will increase across the three AccWalker conditions, and 2) EEG PSD within the delta and theta frequency bands will correlate with the temporal and spatial kinematic variables measured using the AccWalker TARGET protocol.

Methods

Participants

Twenty-four participants enrolled in this study. Participant demographics are presented in Table 6. Participants were recruited through local colleges and the greater community of Greensboro, North Carolina. All participants were excluded if they failed to meet any of the following inclusion criteria: no history of neuromuscular condition, no history of a brain injury resulting in loss of consciousness, no history of a concussion 12 months prior to study participation, no family history of epilepsy, and no current use of psychoactive recreational or prescription drugs. This study was approved by the University of North Carolina at Greensboro's Institutional Review Board.

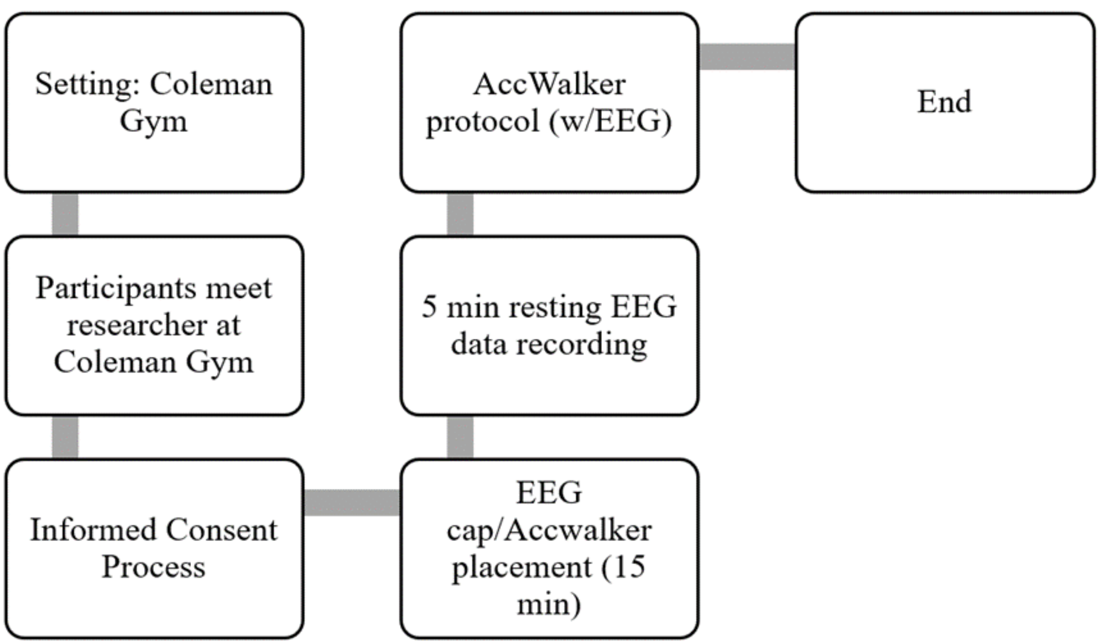
Table 6. Participant Demographics.

Sex	n	Height (cm)	Weight (kg)	Age (yrs)
Female	17	163.6	62.2	22.5
Male	7	173.9	79.5	26.0
Mean		166.6	67.2	23.5
SD		9.3	13.6	4.6

Experimental Design

Participants met with the researcher in the research gymnasium, where the experiment was formally explained, and an informed consent form was signed. After informed consent was given The EEG caps and amplifier were affixed to the head and trunk of the participant (the amplifier communicated with the Brain Vision recorder via Bluetooth and was secured within a chest strap on the posterior of the participant). After, the EEG cap and amplifier were secured an impedance check of all electrodes was performed ensuring that impedance was ≤ 50kΩ. After a five-minute rest period, the AccWalker phone (Google Pixel) was affixed to the lateral side of the participants dominant thigh and the steeping in place movement was demonstrated for the EO condition and a 30s practice of this condition was performed. After this, the participant was asked to perform a 30s practice of the HS condition after a short demonstration. Then six full trials (2 x EO, 2 x EC, 2 x HS), lasting 70s each were performed, with the participant expressing that that were able to continue after each trial; a flowchart of the procedure is in Figure 3.

Figure 3. Data Collection Protocol.



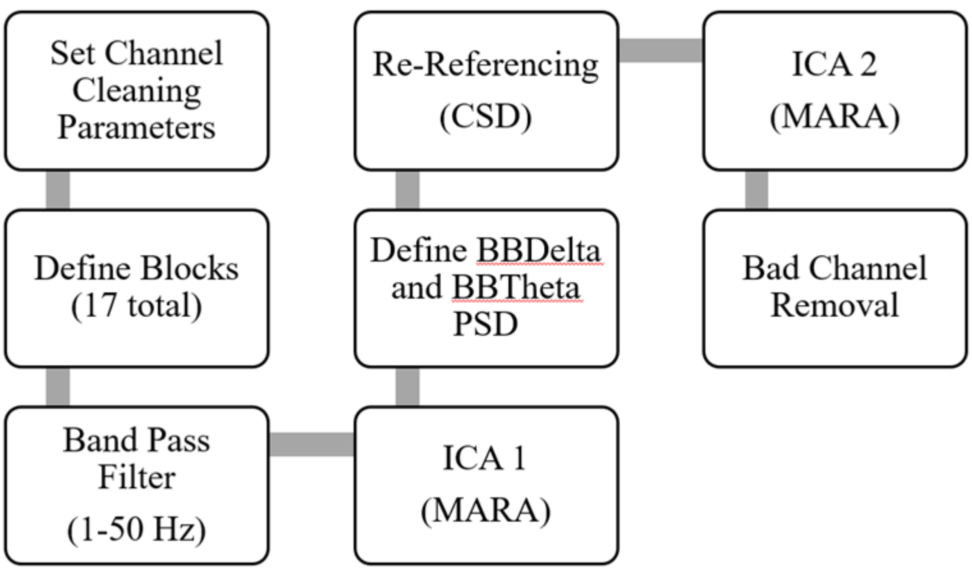
Each Participant performed the six balance tasks of the AccWalker TARGET protocol. AccWalker is a novel Android-based application that allows for the measurement of the movement of a participant's movement and orientation around three axes: pitch, yaw, and roll (recorded at 96 Hz). The app uses a standard Android sensor fusion algorithm called rotation vector to combine data from the phone's accelerometer, gyroscope, and the geomagnetic field sensor to calculate orientation. These data allow for the capture of the orientation of the thigh segment during stepping-in-place and then derive a series of movement parameters such as mean stride time, stride time variability, range of motion of the thigh, and variability of the thigh movement. The full protocol calls for three conditions for stepping in place: with eyes open (EO), with eyes closed (EC), and while moving head left-right, in a "Shaking No" manner (HS). The AccWalker was placed on the lateral thigh of the dominant thigh (participants were asked which leg they lead with when climbing stairs) midway between the center of the hip and knee joint. The entire AccWalker protocol (TARGET) consisted of two practice sessions (each 30 s) of the EO and HS conditions, two trials each (all 70 s) of the EO, EC, and HS conditions. In all conditions, participants stepped in place to a 10 s metronome embedded in the app, after which they were instructed to continue stepping at that pace for the final 60 s of the trial.

EEG data were collected using a 32 electrode R-Net EEG cap (Brain Products, Zeppelinstraße, Germany), sampled at 500Hz, and recorded in Brain Vision Recorder (Brain Vision, Morrisville, NC, USA). Participants sat quietly and as still has possible for 5 minutes before the start of the TARGET protocol. We continuously recorded EEG data at the start of rest and through the entire Target protocol (approximately 15 minutes total). The beginning and end of each TARGET trial was logged by a "1" (Begin) or "2" (End) by a keyboard strike on the computer. Time measurement for the beginning and end of each trial were also logged manually in the event of the keyboard strike being logged incorrectly or omitted.

EEG Preprocessing

Data were imported offline into Matlab version R2020b (MathWorks, Natick, MA, USA) for preprocessing using tools from EEGLAB (version 2021.1) (Delorme & Makeig, 2004) (pop_loadbv). Scripts and wrappers were modified from BeMobileLab (https://github.com/BeMoBIL/bemobil-pipeline, (Klug et al., 2022)). Data were filtered (pop_eegfiltnew) between 1 Hz and 50 Hz. An independent component analysis (MARA, https://irenne.github.io/artifacts/, (Winkler et al., 2011)) was performed to filter non-brain components from the recorded EEG signal (Rejer & Gorski, 2015). Components of non-brain origin were automatically rejected (processMARA, https://github.com/irenne/MARA/blob/master/processMARA.m). Bad channels were identified by the preprocessing parameters: chancorr_crit, Chan_max_broken_time, and Chan_detected_fraction_threshold functions within the BeMobileLab Matlab script. The Chancorr_crit parameter is a part of the clean_artifacts function available a plugin for EEGlab. This parameter represents a correlation threshold in which a channel may be rejected if its correlation to other near (spatially) channels is less than that of the preconfigured value (the default is 0.8) (Klug et al., 2022; Kothe & Makeig, 2013). For this study, due to the dynamic nature of the intervention we selected a lax value of 0.6. The Chan_max_broken_time parameter is a value of a proportion of time in which a channel may be marked as a candidate for rejection before it is rejected from the final output; for our study we selected the default value of 0.5 (Klug et al., 2022). The Chan_detected_fraction_threshold parameter will reject a channel if a channel has been determined “bad” after running ten interactions and being marked by the two former parameters for five of those iterations (as the value which we used - the default is set to 0.5) (EEG Preprocessing, n.d.; Klug et al., 2022). “Bad” channels are then interpolated into the cleaned data set using the bemobil_interp_avref function of the bemobil_process_all_preprocessing Matlab script (EEG Preprocessing, n.d.; Klug et al., 2022). A second independent component analysis was performed to further remove non-brain components after bad channels had been removed and the data were subsequently re-referenced to the common average (Average Ref), (Hu et al., 2018). A flowchart of the procedure is in Figure 4. Power spectral density was extracted from three bilateral frontal electrodes (F3, Fz, and F4) (Hülsdünker et al., 2015; Slobounov et al., 2006).

Figure 4. EEG Preprocessing Pipeline.



AccWalker Preprocessing

The AccWalker data was pre-processed using custom Matlab scripts to extract the aforementioned temporal and spatial variables for each of the three conditions (EO, EC, HS). Per the TARGET protocol, two trials were performed for each condition, which were averaged for reliability purposes. The data were then exported to SPSS (v. 25, IBM, Armonk, NY).

Statistical Analysis

Purpose 1: Delta and Theta PSD were calculated as a percentage of all other band activity (Broadband, BBDelta and BBTheta) (e.g., $BBPSD = \frac{PSD\ Delta}{Delta+Theta+Alpha+Beta+Gamma}$. SPSS version (v. 25, IBM, Armonk, NY) was used for the final analysis of the processed Delta and Theta PSD. Separate, repeated measures ANOVAs (RM-ANOVAs) tested the effect of condition (4 conditions: Rest, Eyes Open (EO), Eyes Closed (EC), and Head Shake (HS)) on PSD in each band (delta, theta). Degrees of freedom were corrected (Hunyh-Feldt) in cases where the assumption of sphericity was violated. Effect sizes are reported as partial-eta squared ($\eta p2$). Significant main effects were decomposed by pairwise comparisons, and Type I error rate was controlled by Bonferroni correction for six comparisons $p - \frac{value}{6}$.

Purpose 2: Before analyzing, the individual values for BBDelta, BBTheta, and the temporal and spatial variables measured from the AccWalker device were z-score transformation to account for individual differences $z = \frac{x - \mu}{\sigma}$ within Microsoft Excel and a Shapiro-Wilk test of normality was performed using SPSS (v. 25, IBM, Armonk, NY). The test of normality indicated that our data may not be normally distributed, so we elected to perform Spearman nonparametric correlations for our correlation analysis. Spearman's rho was calculated between the temporal and spatial variables collected via the AccWalker and BBDelta and BBTheta for each condition.

Results

Descriptive statistics of the temporal and spatial metrics of the three AccWalker conditions can be found in Table 7. There was an effect of condition on BBDelta PSD [$F(1.816,49.019)= 25.912$, $p < 0.001$, $\eta p2 = 0.490$, $\varepsilon=.605$] and BBTheta PSD [$F(2.170,58.590)= 22.455$, $p < 0.001$, $\eta p2 = 0.454$, $\varepsilon=.723$]. BBDelta increased from Rest to EO ($p < 0.001$) and from EC to HS ($p < 0.001$) but did not increase from EO to EC ($p = 0.465$). BBTheta increased from Rest to EO ($p < 0.001$) but did not increase from EO to EC ($p = 1.00$), or from EC to HS ($p = 0.069$).

Table 7. AccWalker Kinematic Data.

		AccWalker Descriptives Pre						AccWalker Descriptives Post					
		EO		EC		HS		EO		EC		HS	
	Unit	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Temporal Metrics													
Mean stride time	s	1.13	0.04	1.12	0.04	1.11	0.05	1.11	0.04	1.11	0.05	1.10	0.05
CV stride time	%	2.30	0.55	2.50	0.52	2.74	0.80	2.21	0.50	2.51	0.67	2.52	0.76
Spatial Metrics													
Peak thigh CV	deg	7.41	2.42	8.88	3.95	8.65	2.76	7.18	1.53	8.27	2.75	8.16	2.43
Thigh RoM	deg	38.41	10.26	35.56	10.78	33.12	10.08	34.85	9.71	31.53	9.16	30.16	8.08
Peak lift vel SD	$deg \cdot s^{-1}$	13.81	3.77	14.45	3.10	14.25	3.15	12.95	3.09	12.94	3.21	13.02	2.30
Peak return vel SD	$deg \cdot s^{-1}$	13.93	3.10	15.13	3.10	14.57	3.28	13.24	2.65	13.67	3.32	13.79	2.72

There were no significant correlations between either the BBDelta or BBTheta frequencies and the AccWalker temporal variables (Stride Time CV or Mean Stride Time). BBDelta was positively correlated with Peak RoM CV, $\rho(22) = .48$, $p < 0.05$ in the EO condition and negatively correlated with Thigh RoM, in the EO condition, $\rho(22) = -.53$, $p < 0.01$, and EC condition, $\rho(22) = .57$, $p < 0.01$. BBTheta was positively correlated with Thigh RoM in the EO condition, $\rho(22) = .57$, $p < 0.01$, and with Peak RoM CV in the HS condition, $\rho(22) = .43$, $p < 0.05$; results of our correlation analysis can be seen in Figures 5-9.

Figure 5. Scatterplot and Spearman's Correlation Coefficient (rho (ρ)) calculated between z-scored BBDelta (y-axis) and AccWalker Measured Peak Thigh RoM CV (x-axis). Significance level defined at $p < 0.05$ (*), n=24.

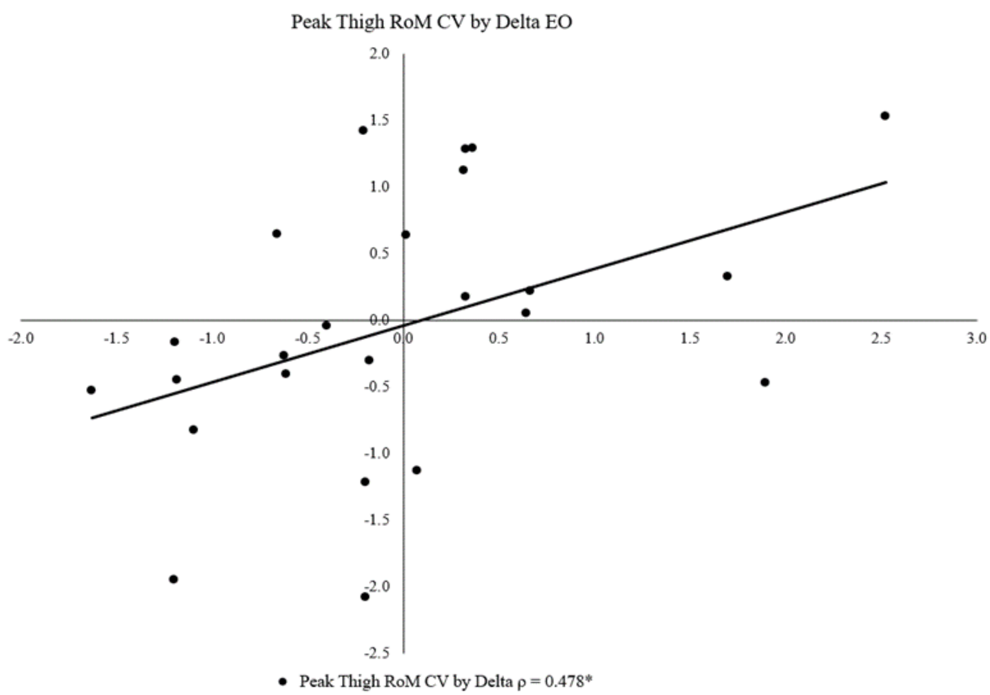


Figure 6. Scatterplot and Spearman's Correlation Coefficient (rho (ρ)) calculated between z-scored BBDelta (y-axis) and AccWalker Measured Thigh RoM (x-axis). Significance level defined at $p < 0.01$ (**), n=24.

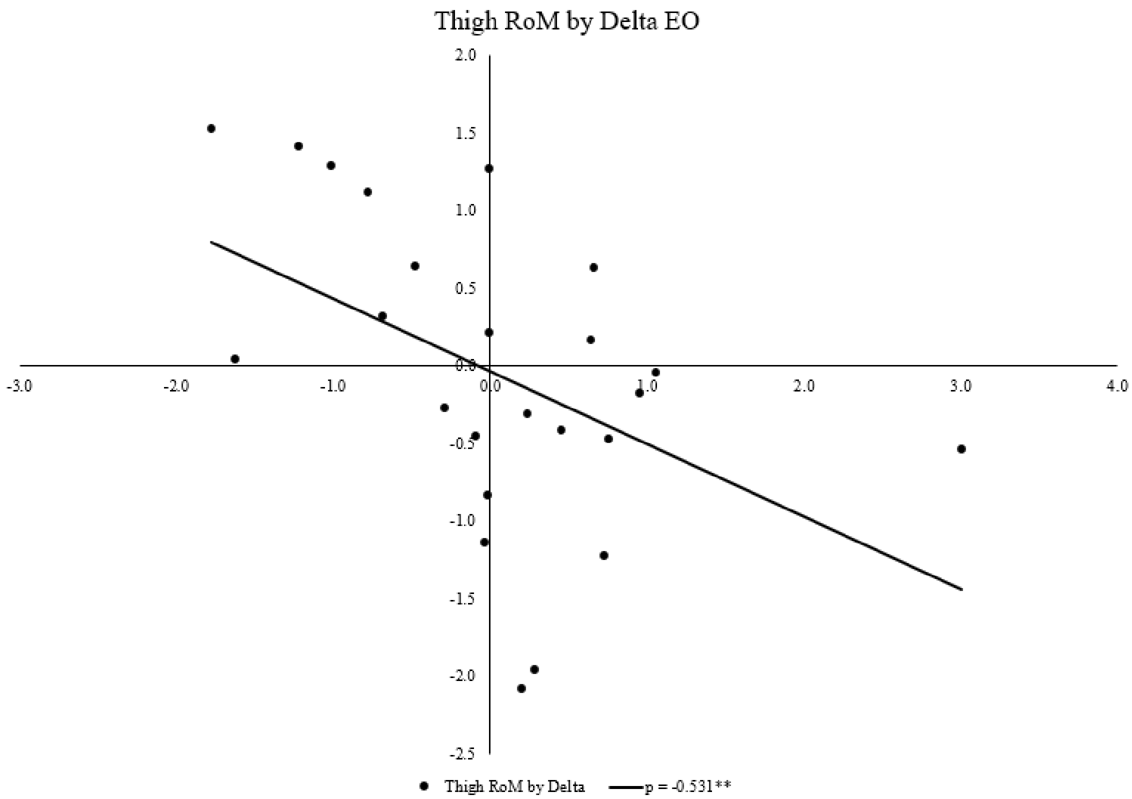


Figure 7. Scatterplot and Spearman's Correlation Coefficient (rho (ρ)) calculated between z-scored BBDelta (y-axis) and AccWalker Measured Thigh RoM (x-axis). Significance level defined at $p < 0.01$ (**), n=24.

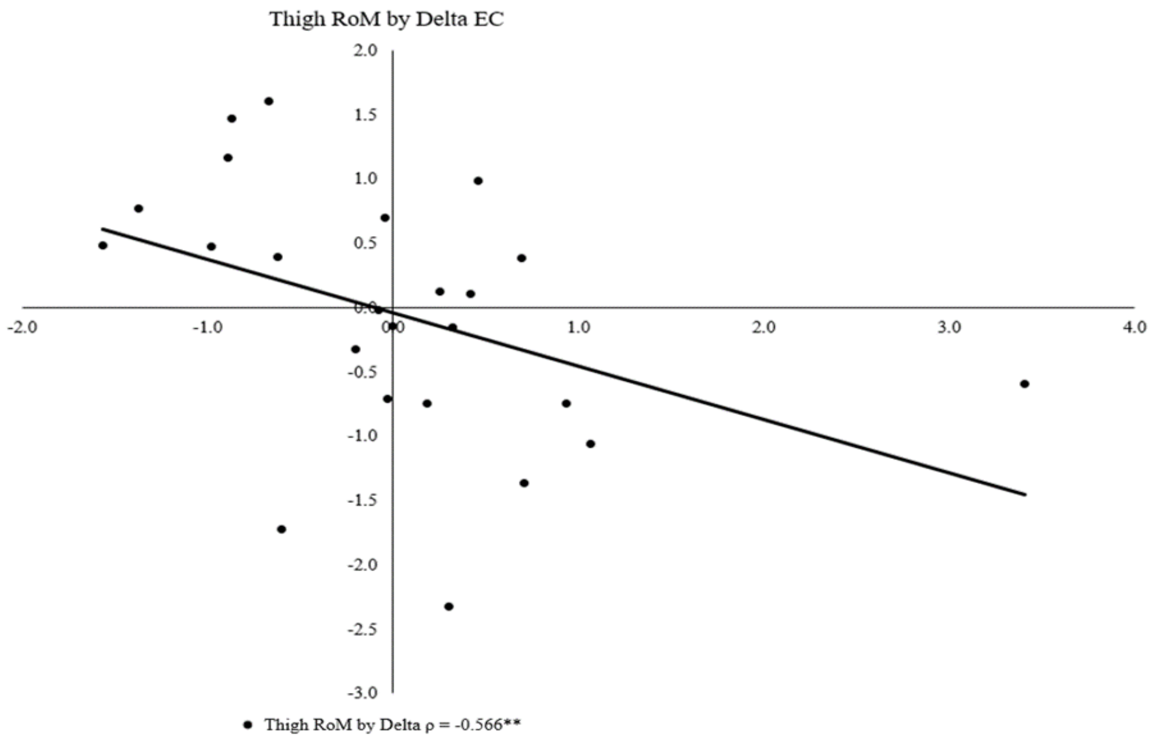


Figure 8. Scatterplot and Spearman's Correlation Coefficient (rho (ρ)) calculated between z-scored BBTheta (y-axis) and AccWalker Measured Thigh RoM (x-axis). Significance level defined at $p < 0.01$ (**), n=24.

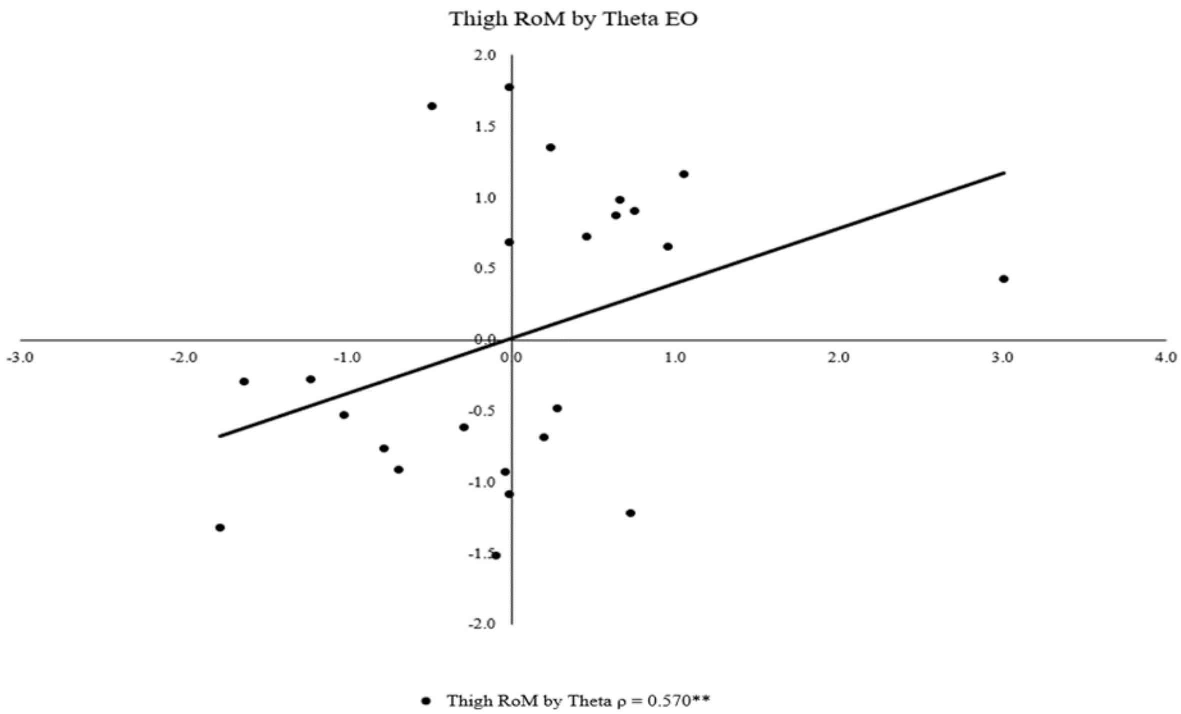
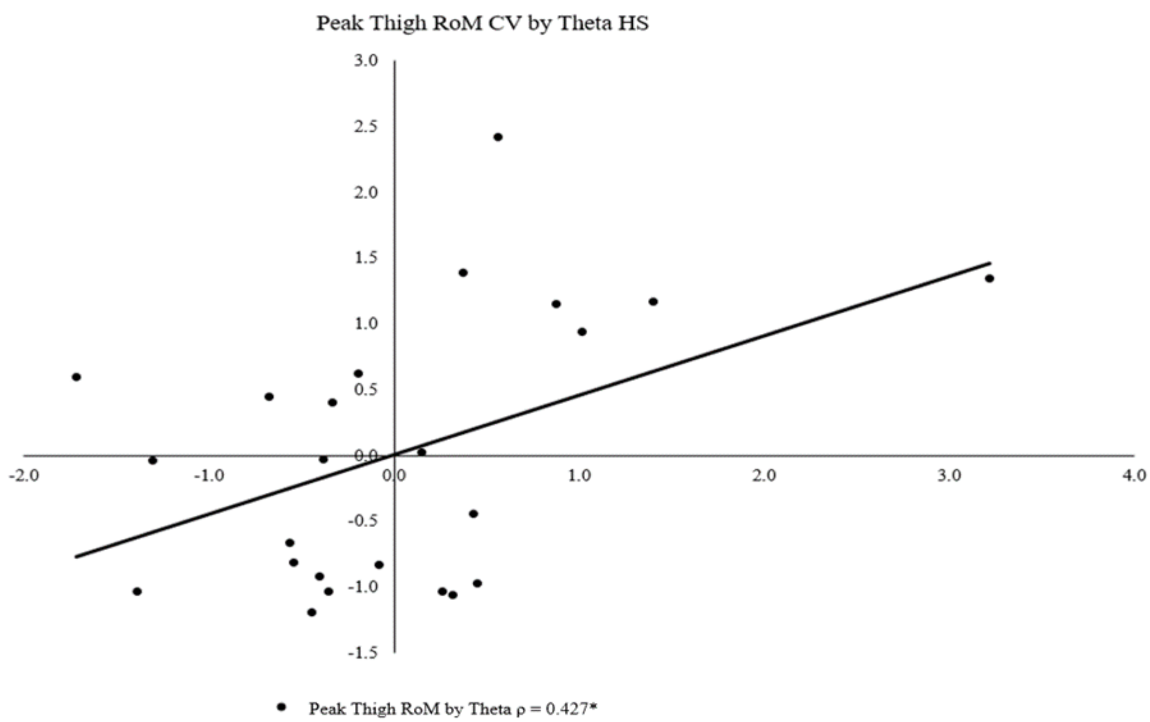


Figure 9. Scatterplot and Spearman's Correlation Coefficient (rho (ρ)) calculated between z-scored BBTheta (y-axis) and AccWalker Measured Peak Thigh RoM CV (x-axis). Significance level defined at $p < 0.05$ (*), n=24.



Discussion

The purpose of study was to compare EEG PSD characteristics of dynamic balance across three different AccWalker conditions, and to examine correlations between EEG PSD characteristics and temporal and spatial kinematic data during a stepping in place task (mTBI Assessment of Readiness Gait Evaluation Test (TARGET)). We had two hypotheses 1) EEG (PSD) within the delta and theta frequency bands will increase across the three AccWalker conditions, and 2) EEG PSD within the delta and theta frequency bands will correlate with the temporal and spatial kinematic variables measured using the AccWalker TARGET protocol. Our results show that both BBDelta PSD and BBTheta PSD increased during the three AccWalker conditions compared to the resting condition. BBTheta PSD did not significantly increase between conditions (EO to EC, EC to HS, or EO to HS), but BBDelta PSD did significantly increase between the EO and HS conditions. Additionally, we found several significant correlations between the AccWalker spatial metrics and BBDelta and BBTheta PSD.

The increase in BBTheta PSD from a stable posture to unstable posture that we observed are consistent with postural instability measured while standing one leg (Slobounov et al., 2009), during balance beam walking (compared to treadmill walking), and during a challenging balance task (i.e., standing on an oscillating platform under varying levels of support) (Hülsdünker et al., 2015; Sipp et al., 2013). Based on prior work using dense EEG arrays and source localization techniques, it is possible to interpret the changes we observed as representing greater recruitment of association cortex, such as the premotor, prefrontal, and anterior cingulate cortex (Sipp et al., 2013; Slobounov et al., 2009) in response to decreased balance stability (Adkin et al., 2006; Hülsdünker et al., 2015). However, our results of no significant increase in BBTheta during the EC condition compared to the EO condition, or the EO to HS, or EC to HS conditions are in contrast to other reports that that perturbations of the visual or proprioceptive sensory systems increases theta activity (Oliveira et al., (2017) and Barollo, et al., (2020). These previous studies suggest that with reduced sensory input (e.g., with the eyes closed or proprioceptive stimulation), greater attentional and/or cognitive resources associated with postulated prefrontal-ACC interactions may be required for successful balance control (Lakatos et al., 2008; Sipp et al., 2013). Our original hypothesis included that as condition difficulty increased (EO < EC < HS), EEG slow rhythm PSD would also increase. This behavior may suggest that the EC and HS (for BBTheta PSD) task demand was not significantly greater than the EO condition. This suggestion may be supported by the kinematic data provided by the AccWalker (see Table 8) where the changes in spatial and temporal kinematic measures, did not appear to be vastly different across the three conditions. However, it may be noteworthy to mention that the participants in the (Oliveira et al., 2017) study were constrained by a safety harness while walking with their eyes closed. While the importance of the safety of research participants cannot be overstated, the authors do note the possibility of the harnessing assisting in the maintenance of their gait and possibly providing increased sensory information (Oliveira et al., 2017).

Our results of an increase in BBDelta PSD from rest to the EO and EC to HS conditions (stable posture to unstable posture), are consistent with postural instability measured with perturbations of the visual and vestibular systems (Aubonnet et al., 2022). These results may suggest that concurrent perturbations of the visual and vestibular sensory systems would require recruitment of cortical motor regions (i.e., prefrontal cortex, premotor cortex) working in concert with the primary motor cortex; thereby, increasing demand upon these interactions and increases in Delta PSD (Ozdemir et al., 2018). Similarly, the HS condition of the TARGET protocol is intended to perturb the sensory systems of the nervous system for the study of how balance may be affected due to forces involving the head (Rhea et al., 2017). The HS condition is similar to the movable platform and virtual environment of a “ship at sea” employed by Aubonnet et al., (2022), and may share in the ability to disrupt the vestibular system via the semicircular canals (Purves et al., 2012); which may indicate not only a response to greater sensory stimulation (Aubonnet et al., 2022; Ibitoye et al., 2021), but also an increase in attentional selection resulting in increased delta band oscillations (Lakatos et al., 2008).

Previous work employing the AccWalker has suggested significant changes in spatial metrics between the conditions of the TARGET protocol, notably significant decreases in thigh RoM from the EC to HS conditions and increases in Peak RoM CV in the HS condition (Rhea et al., 2022); which are similar to our results. Additionally, an effect between concussed and non-concussed participants displayed an increase in the variability of maximum velocity of the thigh while stepping in place for previously concussed participants (Rhea et al., 2022). Our correlation analysis revealed positive associations between variability in Peak RoM CV of the thigh and BBDelta PSD within the EO condition. In other words, as the variability of the achieved height of the thigh increased while stepping in place, delta OSD also increased. This relationship may suggest that within conditions of increased instability, the aforementioned increase in BBDelta may suggest the actions of inhibitory oscillations needed for increases attention during conditions of instability (Aubonnet et al., 2022; Ibitoye et al., 2021; Ozdemir et al., 2018). The negative correlations between Thigh RoM and BBDelta PSD during the EO and EC conditions, represent a similar concept. As in as thigh RoM increased (representing greater stability), BBDelta SD decreased. Further, the positive relationship between Peak Thigh RoM CV and BBTheta PSD during the HS condition is consistent with previous literature, which suggest that during the transition from a stabile to unstable posture (Sipp et al., 2013; Slobounov et al., 2009) or dynamic balance (Hülsdünker et al., 2015) increased activity of between the anterior cingulate cortex (ACC) (Slobounov et al., 2009) (Slobounov et al., 2009), and the prefrontal cortex (Botvinick et al., 2001; Gehring & Knight, 2000). Our result of a positive correlation between Thigh RoM and BBTheta PSD during the EO condition is in contrast with the work if Hülsdünker et al., (2015). However, other work has also no significant increases in theta activity during a balance activity, and have suggested this may be a result of opposing event-related changes in theta power that may not be reflected during a continuous balance tasks (Edwards et al., 2018). Further, the positive correlation of BBTheta PSD and Peak Thigh RoM CV during the HS condition, may suggest that task difficulty may be linked to changes in theta activity (Edwards et al., 2018). These relationships may reflect that with an increase in sensory stimulation (shaking one's head or vibratory stimulation of the proprioceptive system) an up-regulation of cortical activity (Barollo et al., 2020). The lack of correlation of the delta activity to the kinematic variables of the TARGET protocol during the HS condition is contrary to our hypothesis. A reason for this may be because of the task demand of the HS condition. The HS condition is designed to provide for a perturbation of the vestibular system (Rhea et al., 2022). Further, our analyses did show an effect of condition for the delta frequency during the HS condition compared to the EC condition. While this result may be a consequence of noise generated by the rhythmic movement of the head similar to the periodicity of the delta frequency (Castermans et al., 2014), thus, artificially inflating the BBDelta PSD measure. It is also possible that the demand of this condition (Rhea et al., 2023)may have led to a desyncing of head and thigh movement, leading to a disturbed frequency of noise that may have masked the delta frequency and been uncorrelated to thigh movement.

Our study did have limitations however, that indicate that the result of this study should be interpreted with caution. One such limitation may have been that during the EC condition, the participants were only asked to close their eyes and no additional measures were taken to ensure restricted vision (e.g., a blindfold). While the research staff did ensure that the participants' eyes remained closed during the entirety of the EC condition, we cannot be sure that lighting in the gymnasium or slightly closed eyes did not provide some additional sensory input. This additional sensory input could have limited the validity of the eyes closed condition if participants were to be exposed to information of their environment for a brief moment, therefore not allowing for a true vision – no vision comparison (Oliveira et al., 2017). However, while no differences in BBTheta PSD from EO to EC or HS were seen, a significant increase in BBDelta PSD was seen in HS compared to EO; this may suggest that steeping in place while shaking one's head may increase the instability of the participants' balance and thus an increase in delta power (Ibitoye et al., 2021). Further, a similar finding by Aubonnet et al., in 2022 was reasoned to be a consequence of constant unstable balance (Aubonnet et al., 2022). BBDelta PSD increase could also have also been the result of a polluted signal within the Delta frequency range originating from low-frequency rhythmic head movements (Castermans et al., 2014). Specifically, the head movement required by the HS condition is within the Delta band could be disproportionately increasing delta power (Castermans et al., 2014).

In conclusion, this study has expanded on previous literature indicating increased involvement of the frontal-central and central regions of the brain during perturbed balance. Additionally, the TARGET protocol allowed for the examination of neural correlates under 3 different sensory conditions, which again may allow for increased external validity to lab-based measures. This is supported by an increase in Delta and Theta activity seen during a dynamic and continuous balance task and the correlation of the former activity to kinematic variables of dynamic balance. Further, while previous work has examined similar EEG characteristics along with balance performance measures, these previous measures where predominantly of a non-dynamic nature. However, this study expanded on previous work with the inclusion of a dynamic balance task which can allow for tri-planar movement, which may allow for a better simulation of "real world" tasks, which are generally of a dynamic in nature (Basford et al., 2003; Chou et al., 2004; Guskiewicz, 2001; Kuznetsov et al., 2018; Rhea et al., 2017). Finally, this study expands upon the simultaneous use of EEG and balance assessment, specifically as it is the first study to use a truly dynamic balance task along with a 32-electrode mobile EEG system. The TARGET protocol combined with EEG may be important for continued study of unaffected balance and neural changes due to injury (Monroe et al., 2020; Rhea et al., 2017) or pathological processes (Goble et al., 2014).

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APPENDIX A: INFORMED CONSENT FORM

UNIVERSITY OF NORTH CAROLINA AT GREENSBORO

CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM

Project Title: Effects of Repetitive Sub-concussive Head Impacts on Postural Control and Electro-Cortical Activity

Principal Investigator: John Palazzolo, M.S.
Faculty Advisor: Christopher K. Rhea, Ph.D.

Participant's Name: ____

What are some general things you should know about research studies?
You are being asked to take part in a research study. Your participation in the study is voluntary. You may choose not to join, or you may withdraw your consent to be in the study, for any reason, without penalty.

Research studies are designed to obtain new knowledge. This new information may help people in the future. There may not be any direct benefit to you for being in the research study. There also may be risks to being in research studies. If you choose not to be in the study or leave the study before it is done, it will not affect your relationship with the researcher or the University of North Carolina at Greensboro.

Details about this study are discussed in this consent form. It is important that you understand this information so that you can make an informed choice about being in this research study.

You will be given a copy of this consent form. If you have any questions about this study at any time, you should ask the researchers named in this consent form. Their contact information is below.

What this study is about?
This is a research project. Your participation is voluntary. The goal of this study is to examine your balance, brain activity, heart rate, before and after kicking 10 soccer balls over 10 minutes.

Why are you asking me?
You must also be able to maintain an upright stance for up to an hour and be able to walk-in-place unaided for up to 10 minutes. You should not participate if you have any current neuromuscular injury/disorder that affects gait or balance (as well as, not having suffered a concussive event within the past 12 months) and activities of daily living, visual impairment not correctable with lenses, abnormal balance function, surgery within the past 6 months have had a brain injury resulting in loss of consciousness, have hearing difficulties, have a family history of epilepsy, have a history of cardiovascular or respiratory disease, are currently taking prescribed or recreational any psychoactive drugs, or use of lower extremity assistive device or prosthetic device. You must be between the ages of 18-50 to participate.

What will you ask me to do if I agree to be in the study?
You will be asked to partake in the following events over an approximate 105 minute:

This study will measure balance, brain activity, and cardiovascular activity before and after kicking a soccer ball.

A list of the assessments and soccer task is below:

At the beginning of your participation, you will be asked to complete a history questionnaire, to determine if you meet study inclusion criteria.

AccWalker (balance tests): The AccWalker is an investigational device. You will have a strap placed around your upper thigh. This will. contain a smartphone. You will be asked to perform 9 tests. These tests will require you to step in place to a continuous rhythm. While stepping in place there will be one test with your eyes open, one with your eyes closed, and one with your eyes open but you will be shaking your head (shaking "No") sideways. The entire procedure will take approximately 10 minutes and consist of two practices, and three trials.

Electroencephalography (EEG): A head cap with 32 electrodes will be placed on your head along with a body harness to secure the EEG device. During each of the balance tests, your brain activity will be recorded.

Heart Rate monitoring: You will have a heart rate monitor attached (via an elastic strap) placed against your chest. This device will monitor your heart activity during each of the balance tests.

Soccer Task: You will be asked to complete a soccer ball kicking task, 1) you will be asked to kick a soccer ball (Size 5 soccer ball, 400 g, 70 cm circumference, inflated to 12 psi, launched at a velocity of 10.75 kilometers per hour, (approximately 24 miles per hour) that is launched towards you from a soccer ball launching machine. You will be asked to complete 10 kicks in 10 minutes.

Schedule

The following is a chronological list of the procedural steps for data collection:

1.Participants meet with the researcher at the Coleman Research Gym (Coleman 248).
2. Informed consent process (15 min)
    a. After informed consent process, participant will complete a demographics (sex, age, height, and weight), and injury history questionnaire to determine if they meet study inclusion criteria (please see attached document).
3. Measurement device placement period (approximately 15 min)
    a. AccWalker strapped to the thigh
    b. A heart rate monitor will be strapped across the participant's chest
    c. Participant will be seated, and a pre-soaked EEG cap will be placed on the participant's head.
4. EEG harness attached to the back of the participant
    a. Saline will be added, and adjustments made to reduce impedance at the electrodes of the EEG cap
5. 5-minute resting EEG data recording
6. Pre-soccer kicking task AccWalker, EEG, HRV assessment (approximately 10 min)
8. 10 min break in which, all devices are removed.
9. Soccer kicking task
10. Measurement device placement period (approximately 15 min)
    a. AccWalker strapped to the thigh
    b. A heart rate monitor will be strapped across the participant's chest
    c. Participant will be seated, and a pre-soaked EEG cap will be placed on the participant's head
11. EEG harness attached to the back of the participant
    a. Saline will be added, and adjustments made to reduce impedance at the electrodes of the EEG cap.
12. 5-minute resting EEG data recording
13. Post-soccer kicking task AccWalker, EEG, HRV assessment (approximately 10 min)
14. End of study participation

Your length of involvement will be one visit lasting approximately 105 min for balance, EEG, heart rate, and the soccer task.

You may stop the study at any time, for any reason.

Is there any audio/video recording?
There will be no video or audio recording during the testing session.

What are the dangers to me?
The Institutional Review Board at the University of North Carolina at Greensboro has determined that participation in this study poses minimal risk to participants. There is minimal risk that you could experience discomfort from stepping-in-place for 8 tests of 30 to 70 seconds. Musculoskeletal injury from falling (from dizziness due to rotation of the head), or impact with other equipment or furnishings (due to having your eyes closed); the research personnel will actively ensure that the participant is at a safe distance from any obstacles and is comfortable with continuing participation between tests. To ensure safety during the soccer ball kicking protocol, you will be asked if you are ready to continue with the next ball or if you would like to stop in between each soccer ball kick attempt.

If you have questions, want more information, or have suggestions, please contact John Palazzolo at j_palazz@uncg.edu or Christopher Rhea at ckrhea@uncg.edu. If you have any concerns about your rights, how you are being treated, concerns or complaints about this project or benefits or risks associated with being in this study please contact the Office of Research Integrity at UNCG toll-free at (855)-251-2351.

What if I get injured?
UNCG is not able to offer financial compensation nor to absorb the costs of medical treatment should you be injured as a result of participating in this research study. However, we will provide you with a referral to student health or your primary care physician. You do not waive your legal rights by signing this consent form.

Are there any benefits to me for taking part in this research study?
There are no direct benefits to you for participating in this study.

Are there any benefits to society as a result of me taking part in this research?
The results of this project may inform basic and clinical science about the effects of low-level (sub-concussive) impacts on postural control. Along with a more complete understanding of nervous system control in balance and possible disruption of this control from sub concussive exposure.

Will I get paid for being in the study? Will it cost me anything?
There are no costs to you for participating in this study. After your participation, a $10 Target gift card upon completion of the study. Additionally, upon completion of the study, all participants will be entered into a drawing to receive one (of 9) additional $20 Target gift cards or one (of 5) $50 Amazon gift cards.

How will you keep my information confidential?
All information that is obtained from this study is strictly confidential unless disclosure is required by law. All data (written and electronic) will only contain your assigned code number. The list connecting your name to your assigned code number will be kept in a locked file cabinet within a locked office in the VEAR laboratory separate from all data. The VEAR laboratory is protected by an intellikey. All consent forms will be maintained in a confidential file only accessible by the investigator and faculty advisor. When the study is completed and the data have been analyzed, this list will be destroyed. Your name will not be used in any report. The consent forms will be kept in a file in a locked room for three years at

which time they will be destroyed by shredding. All data will be stored on the principal investigator's personal computer identified only by subject number. All data disks will be erased once all manuscripts of the data have been submitted and published for two years. A photocopy of this original consent form will be provided to you for your records. The FDA has the right to inspect the research records and could possibly inspect records associated with this study.

Will my de-identified data be used in future studies?
Your data will be destroyed 5 years after the submission and publication of any relevant manuscript. De-identified data will not be stored and will not be used in future research projects.

What if I want to leave the study?
You have the right to refuse to participate or to withdraw at any time without penalty. If you choose to withdraw, it will not affect you in any way, and you may request that any of your data which has been collected be destroyed (unless it is in a de-identifiable state). The investigators also have the right to stop your participation at any time. This could be because you have had an unexpected reaction, or have failed to follow instructions, or because the entire study has been stopped.

What about new information/changes in the study?
If significant new information relating to the study becomes available which may relate to your willingness to continue to participate, this information will be provided to you.

Voluntary Consent by Participant:
By signing this consent form, you are agreeing that you have read, or it has been read to you, and you fully understand the contents of this document and are openly willing to take part in this study. You are also confirming that all of your questions concerning this study have been answered. By signing this form, you are agreeing that you are 18 years of age or older and are agreeing to participate, in this study described to you by ______.

Signature: ____ *Date:* ________________________

APPENDIX B: DEMOGRAPHICS AND INJURY HISTROY FORM

Effects of Repetitive Sub-concussive Head Impacts on Postural Control and Electro-Cortical Activity Subject Demographics

Sex____________________

Age____________________

Height (cm)______________

Mass (kg)_______________

PHYSICAL ACTIVITY AND HEALTH HISTORY

Do you have any General Health Problems or Illnesses? (e.g., diabetes, respiratory disease)
Yes____ No____ if so, please state: -

____

Do you have any history of any conditions which may affect your balance? (e.g., Concussion within the last year, Multiple Sclerosis, inner ear disorders) Yes____ No____

Please list any previous injuries to your lower extremities. Please include a description of the injury (e.g., ligament sprain, muscle strain), severity of the injury, date of the injury, and whether it was on the left or right side.

Body Part Description Severity Date of Injury L or R

Hip

____

____

Thigh

____

____

Knee

____

_____

Lower Leg

____

_____

Ankle

____

____

Foot

____

____

Please list any previous surgery to your lower extremities (Include a description of the surgery, the date of the surgery, and whether it was on the left or right side)

Body Part________**Description**__________________**Date of Surgery**__________

L or R

____

____

____

____

Please answer the following questions:

What year did you start playing soccer? ___________________________

Have you played consistently since you stated? yes_____, no_____

If No, why did you stop and/or what were your gap years

____

____

____

What was your completion level (e.g.,) recreational league, school league, university club level, university intramurals, etc.); and what ages did you participate at those levels

____

____

____

____

Approximately how many games did you play per year? If it varied by age or competition level, please indicate that.

Approximately how many practices did you participate in per year? If it varied by age or competition level, please indicate that.

What was your position(s)? If it varied by age or competition level, please indicate that.

Do you recall at what age or level you started heading the ball in practice or games?

What is your best estimation of how many times you headed the ball per year? If it varied by age or competition level, please indicate that.

Have you participated in other sports where head contact was involved (e.g., football, hockey, field hockey, etc.)? If so, please indicate the sport(s) and years you participated in those sports.

APPENDIX C: SPSS OUPUT FOR TEST OF NORMALITY FOR ACCWALKER

Z-TRANSFORMED METRICS

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
ZAW_STCV_EOA	.138	24	.200*	.926	24	.080
ZAW_STCV_ECA	.144	24	.200*	.893	24	.015
ZAW_STCV_HSA	.168	24	.076	.911	24	.037
ZAW_MST_EOA	.177	24	.049	.952	24	.307
ZAW_MST_ECA	.128	24	.200*	.950	24	.267
ZAW_MST_HSA	.112	24	.200*	.977	24	.835
ZAW_PKRoMCV_EOA	.144	24	.200*	.937	24	.137
ZAW_PKRoMCV_ECA	.215	24	.006	.782	24	.000
ZAW_PKRoMCV_HSA	.132	24	.200*	.917	24	.050
ZAW_TGHRoM_EOA	.115	24	.200*	.938	24	.147
ZAW_TGHRoM_ECA	.128	24	.200*	.890	24	.013
ZAW_TGHRoM_HSA	.080	24	.200*	.965	24	.548

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction