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Attentional focus is the direction of attention during a specific task which can be divided into an external focus of attention (EFA) and an internal focus of attention (IFA). An external focus has been found to be more effective in motor learning and control. Predominately used in sports environments, especially running and gymnastics, it has recently become more utilized in rehabilitation environments. A primary concern with aging is the maintenance of balance, research has consistently shown that balance begins to deteriorate faster after age 65, which in turn leads to injuries and reduced quality of life. Using attentional focus as a strategy that both therapists and individuals can use assists with furthering the study of movement behavior with respect to enhancing and maintaining quality of life.

The purpose of this study was to further examine the influence of attentional focus (EFA and IFA) and its underlying mechanisms in the brain using electroencephalography (EEG) during a balance task. Individuals who are healthy and without neurological impairment were recruited (n=32). A between-subjects design was used to examine potential cortical activation differences between IFA and EFA using the standardized BESS balance protocol. The BESS protocol consists of six tasks all with eyes closed on a firm surface and on a foam surface. Cortical activity measured in the Alpha and Beta bands was examined in effort to investigate differences during attentional focus conditions. This study found postural sway to be reduced in the IFA condition. This study also saw a reduction in cognitive effort in the EFA group. In the frontal region, EFA had decreased alpha, and increased beta power. In the medial region, it was found that there was a broad decrease in mean alpha, as the balance performance decreased. While the beta power broadly increased as balance performance improved. This demonstrates the potential for cortical processing differences when focus is manipulated during a balance task.

ATTENTIONAL FOCUS AND BALANCE CONTROL USING ELECTROENCEPHALOGRAPHY

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Sydney E. Rossback

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Approved by

____
Dr. Louisa D. Raisbeck
Committee Chair

APPROVAL PAGE

This thesis written by Sydney E. Rossback has been approved by the following committee of the Faculty of The Graduate School at The University of North Carolina at Greensboro.

Committee Chair ____________________________
Dr. Louisa Raisbeck

Committee Members ____________________________
Dr. Christopher Rhea

____
Dr. Derek Monroe

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Date of Acceptance by Committee

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Date of Final Oral Examination

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CHAPTER I: INTRODUCTION

Maintaining functional independence is important as we age, with a primary focus on remaining balance (Vaughan et al., 2016). The rate of falls increases after the age of 65, with 25 percent of older adults experiencing at least one fall (CDC, 2017). Falls in older adults can lead to decrease in physical fitness contributing to the onset of chronic diseases and leading to increased costs of healthcare. Therefore, it is important to identify preventative strategies that can alleviate the healthcare system in addition to cost for older adults (Haddad et al., 2019). Balance interventions are inconclusive with their success rates, making it difficult to identify a strategy that can be generalized across a population of older adults. Although we understand the benefits of incorporating different motor behavior strategies into applied environments, such as exercise and rehabilitation programs, there is still a disconnect between what the research shows and how applications occur (Sugiyama & Liew, 2016). Incorporating motor behavior strategies into applied settings has the potential to develop more effective learning and performance over time. More effective balance strategies could lead to more effective movement processing thus improving or maintaining balance over time (Higgins et al. 2021). Current balance interventions primarily focus on functional lower body strength and have shown to decrease fall risk for both men and women (Rubenstein et al, 2000, Judge et al, 1993), Until recently the longest balance intervention was 12 weeks, however Higgins et al (2021), completed a 20 week intervention of 12 weeks training and then 8 weeks of follow up testing (Higgins et al., 2021). As fall rate increases over time, it becomes increasingly important to minimize age-related declines in balance control by verifying potential strategies.

One motor behavior theory that has gained traction over the past two decades is attentional focus. It has changed the way practitioners are delivering instructions to enhance learning and performance (Raisbeck et al., 2020). When instructional cues are delivered, they direct an individual's attention to either and internal or external component to the action (Wulf et al., 2001). Attentional Focus is used by changing the direction of attention toward a specific detail about the movement (IFA) or environment (EFA) (Wulf et al., 2001). Initial studies examined attentional focus on task performance and found improved static balance under EFA conditions (McNevin & Wulf, 2002). Current training studies have demonstrated that integrating attentional focus into the training has shown to be a facilitative tool when training motor skills. Specifically, using an external focus of attention has shown to improve performance (Kal et al., 2013; Wulf et al., 2001). This has been studied in multiple populations and across differing amounts of time. However, it is relative unknown how neurological mechanisms interact and the activation patterns that are involved in the attentional focus and balance practice. The field has continued to grow investigating sports rehabilitation and more recently neuromechanics. The research is consistent in that it has shown adopting an EFA is widely accepted as the more effective method however we know little are the underlying neurological reasons as to why. Understanding the organization and activity patterns in the brain during movement processing can lead to improvements in motor learning. More effective movement processing is the goal for most balance rehabilitation programs. Instructional cues and attentional focus can be applied in practice to facilitate functional independence for individuals. Further work is needed to better understand the interaction of attentional focus during balance from a neuromechanics point of view. The current study used electroencephalography (EEG) to examine cortical activity in young, healthy individuals during a challenging balance task. The purpose of this study was to identify meaningful cortical patterns underlying attentional focus during balance using EEG.

CHAPTER II: REVIEW OF LITERATURE

Overview

The purpose of this review of literature is to examine how attentional focus and balance is reflected in cortical activation, in addition to instructional cueing and the implications instructional cueing has on motor performance and learning. Additionally, this literature review will concentrate on attentional focus, specifically using EEG, during a balance task, and identify cortical activity relationships for internal and external focus of attention instructions.

Attentional Focus Theories and Motor Performance

Attention is cognitive and behavioral; it includes concentration on an aspect of observation (Lindsay, 2020). A theory that has moved to the forefront of research in motor behavior is attentional focus, which suggests that if individual's direct their attention in a specific way that it changes the trajectory of their performance and learning (Wulf et al., 1998). Attentional focus is the direction of attention during a specific task, it has been broken down into external focus of attention and internal focus of attention (Wulf et al., 1998). Internal Focus of Attention (IFA) is when attention is directed inwards to the movement of a person's body (Wulf et al., 1998). External Focus of Attention (EFA) is when attention is directed toward the effect that the movement has on the environment (Wulf et al., 1998). Using specific words (instructional cues), it has been observed that individuals will direct their attention either internally or externally impacting their overall performance and learning.

Over the past several decades the attentional focus literature has consistently shown that using an external focus of attention for most motor tasks will lead to superior performance (Wulf et al., 1998; Vaz et al., 2019). These findings can be explained by the Constrained Action Hypothesis (CAH) suggesting that when you constrain a system, it becomes difficult to be automatic with movements that are typically without conscious thought (Wulf et al., 2001). Using an EFA is thought to encourage self-organization and thus promotes automaticity (Vidal et al., 2018; Kal et al., 20013; Wulf et al., 2001).

Attentional focus has shown to be an effective training tool across a multitude of disciplines from athletes (Vaz et al, 2019; Wulf, 2008), therapy patients and clinical populations. Chronic stroke (G. Kim et al., 2017), Multiple Sclerosis (Shafizadeh et al., 2013), Parkinson's Disease (Canning, 2005; Landers et al., 2005), and even sleep-deprivation (Diekfuss et al., 2018). In the study by Vaz et al. (2019) researchers used a balance board to challenge coordination of 22 young, healthy adults while each participant was given either external or internal focus instructions. The main performance using multiscale entropy, results showed that balance performance increased in the EFA group with lower entropy, meaning their movement was more constant and predictable (Vaz et al., 2019). When we see an increase in entropy in the IFA group, could be explained by the CAH, in that there was inhibition of movement coordination due to hyperconsciousness of the body. Wulf (2008) used attentional focus strategies to examine elite acrobats during a balance task to see if AF effects were generalizable. Using a balance disc, COP was collected for all subjects across all conditions. Performance differences were not seen between the EFA and IFA groups, however, there was an increased performance in the control group (Wulf, 2008). Results suggest that elite athletes have fine-tuned their movement control so correcting errors becomes reflexive (Wulf, 2008). So, using the "reflexive" method of focus, led to the best performance outcome when studying experts (Wulf, 2008).

Attentional focus strategies have been used in clinical populations Kim et al. (2017) used attentional focus strategies on stroke patients to train their upper extremities with robotics.

Although significant difference between EFA and IFA were not observed, it was suggested that stroke patients learn motor skills differently than a healthy person, so this likely has a large effect on the effectiveness of attentional focus strategies (Kim et al., 2017). Another study examined walking gait in people with Multiple Sclerosis and used focus of attention training during rehabilitation over an 8-week period (Shafizadeh et al. 2013). Participants walked on a treadmill for conditions: baseline, IFA, and EFA. Researchers found an increase in stride length and well as stride speed when under the EFA condition, supporting the potential for rehabilitating those with movement and/or neurological disorders using attentional focus strategies (Shafizadeh et al. 2013). Canning (2005) used a dual-task methodology to examine attentional focus effects for those with Parkinson's Disease. Participants completed a walking task in addition to balancing a tray with cups on. For the task they were either asked to focus on the tray (EFA) or their walking (IFA). In contrast to most attentional focus literature, when participants in this study were given EFA instructions they had poorer performance when compared to IFA conditions (Canning, 2005). Walking is a form of dynamic balance, which may have different implications when it comes to attentional focus than static balance. The study by Landers et al. (2005) also looked at patients with Parkinson's Disease, specifically with a history of falls. Each participant did three different balance trials each under 3 modes of attentional focus: control, IFA, and EFA. The balance conditions consisted of two trials on a firm surface, 1 trial with eyes open, 1 with eyes closed. Then the third was a moving surface, with eyes open. The control and IFA conditions were about equal in terms of balance equilibrium performance in Fallers. However, the EFA condition during the most difficult balance task produced significant reductions in sway measurements (Landers et al., 2005). Another interested finding was that during the third, and most difficult balance task, no one fell under the EFA condition (Landers et al., 2005). But multiple people fell during the control and IFA conditions (Landers et al., 2005). Therefore, this study also showed how EFA can reduce the occurrence of falls in Parkinson's Disease patients. Diekfuss et al. (2018) examined attentional focus differences during a balance task when sleep was deprived. A dynamic balance board was used to challenge balance, during different attentional focus cues. Results showed an increase in balance performance for the EFA condition, and no differences for the IFA or control. When sleep deprived, balance declines naturally, so changing instructions, even slightly, can make a difference in balance performance.

Attentional focus has also been notably used during balance training (Wulf et al., 2004; de Bruin et al., 2009; Richer et al., 2017; Diekfuss, 2017; Kal et al., 2019; Higgins et al., 2021; T. Kim et al., 2017; de Melker Worms et al., 2017; Mak et al., 2020; Pinto et al., 2021; Park et al., 2015; Kupper et al., 2020). Many of these authors concluded that EFA lead to better postural control (Diekfuss, 2017; T. Kim et al., 2017; Park et al., 2015; Richer et al., 2017; Wulf et al., 2004). Diekfuss et al (2017), used a 7-day balance training on a wobble board and used fMRI and DTI pre- and post-intervention (Diekfuss et al., 2017). Results showed that the EFA group had better postural control during performance and retention (Diekfuss et al, 2017). Kim et al (2017) completed a systematic review examining AF effects in motor learning and balance. Results showed that EFA led to better balance during the learning phase of the motor skill Park et al (2015), determined, from 18 articles, that 15 suggested effective outcomes when using EFA (Park et al., 2015). Richer et al (2017) studied attentional focus and cognitive task effects on postural control participants completed a static balance task as well as two cognitive tasks using sway velocity, sway frequency, and muscle activation at the ankle as the main outcome variables. Results showed that EFA reduced postural sway compared to baseline and IFA. Wulf et al (2004), examined young adults and the influence of attentional focus during a postural task

(Wulf et al., 2004). Standing on a balance disc, participants would balance with and without a pole held horizontally in the hands for both EFA and IFA. Results showed reduced postural sway, compared to IFA, in both tasks (Wulf et al., 2004).

Although the research is consistent with showing EFA to be superior, research exists to show that there are no differences between the different attentional focus strategies for; gait stability (Mak et al., 2018), proximal and distal levels of AF during balance (Kupper at al., 2020), ; stroke patients and motor skill improvements using attentional focus instructions (Kal et al., 2019), older adults and the attention focus effects on balance disruption and recovery (De Melker Worms et al., 2017) & dynamic balance skills (De Bruin et al., 2009).

Additionally, there are articles that suggest EFA doesn't have a benefit for mobility and/or balance (Pinto et al., 2021; Higgins et al., 2021). Pinto et al (2021) considered older adults as well as younger adults in a study about the effect that attentional focus has on ADLs for the elderly (Pinto et al., 2021). The participants did a sit-to-stand and stand-to-sit tasks while holding varying levels of liquid filled cups (Pinto et al., 2021). The researchers suggested EFA might not be of benefit in younger or older adults in the mobility needed for ADLs (Pinto et al., 2021). Lastly, an article by Higgins et al (2021), examined postural control training in older adults who have a history of falls, using attentional focus strategies (Higgins et al., 2021). This study consisted of a 12-week wobble board training program with EFA or IFA instructions (Higgins et al., 2021). The researchers found that performance did not increase in the EFA group (Higgins et al., 2021). Although the literature is robust, few studies have further explored attentional focus from a mechanistic perspective.

EEG as a Measurement to Understand Cortical Activity during Attentional Focus Cues

Although, Electroencephalography (EEG) is common neuroimaging technique used in clinical practice as a diagnostic tool for neurological disorders and diseases (Murdoch, 1982). Attentional focus is a relatively new field of research, comparatively, in addition to rarely being used in EEG studies. Therefore, the literature on the relationship between attentional focus and neural activity is limited. Most of the relative literature is fMRI research. Although we understand that MRI provides a holistic approach to understanding what occurs in the brain during specific activities, it is restrictive in that those activities must be performed in the prone position with limited movement. This static structure of MRI is limiting when balance and related mechanisms are being studied. Another approach to examining brain activation during motor movement is by using EEG. Electroencephalography allows for observation of neural oscillatory activity (Crosson et al., 2010). This activity is large groups of neurons firing in the cortex which produces repetitive patterns and display the dynamics of the post-synaptic potentials in the brain. One important aspect of oscillatory activity is that includes the factor of time. Therefore, EEG has strong temporal resolution in comparison to other neural imaging techniques (Crosson et al., 2010). Another important factor of EEG is its considerable cost benefit when compared to other neuroimaging techniques like functional magnetic resonance imaging (fMRI) (Crosson et al., 2010). The cost makes EEG is much more accessible to researchers.

There are multiple frequency bands in EEG signal; Delta, Theta, Alpha, Beta, Gamma. The alpha band (8-12 Hz) was previously thought to reflect a visual rhythm. It is consistently seen that alpha band amplitude increases when the eyes are closed, and decreases when the eyes are open (Klimesch, 2012). This remains true even when the individual is in a completely dark room, which suggests that Alpha band activity is inversely related to cortical activity (Lindgren et al., 1998). More recently, a review on EEG and alpha-band relations reported that alpha frequencies were generally more reactive, as well as have a correlation to attention and information processing (Klimesch, 2012). When studying attention, the frontal region of the brain is often examined because it is thought to be correlated to executive functioning, including attention and memory (Fingelkurts et al., 2003). The circuits that control attention overlap strongly and is now accepted as the frontal-parietal attentional network (Corbetta & Shulman, 2002). For this reason, this study will be examining the Alpha band in the frontal region of the brain in attempt to get a well-rounded view on cortical activity during attentional modification.

The beta band (12-30 Hz) is often studied during voluntary movement which will be beneficial in studying differences and similarities in EFA and IFA during a motor task. Research by Peterson and Ferris (2018) examined theta and beta frequency bands during standing balance and walking using EEG in addition to Virtual Reality (Peterson & Ferris, 2018). Results showed increased beta power in the supplementary motor area during a perturbed balance task (Peterson & Ferris, 2018). They also suggested that the analysis of these power variations provide appreciable observation for the underlying mechanisms during sensorimotor tasks. The beta band is seen to increase during steady state contractions in voluntary movement, in humans and other primates (Baker, 2007; Sanes & Donoghue, 1993). This was hypothesized to be related to the beta band being involved with the processing of proprioceptive feedback (Baker, 2007). There has also been literature which has presented elements of voluntary balance tasks involving the primary motor cortex (Beloozerova et al., 2003; Bolton et al., 2012; Jacobs & Horak, 2007; Taube et al., 2006; Baker, 2007; Sanes & Donoghue, 1993). For EEG specifically, the data is taken from the scalp, which gives a surface level view of cortical activity. Using EEG, we can look at general areas to give insight as to the activities that region is possibly involved in.

The role of EEG in balance training is increasingly important. Although balance is widely studied kinematically, the neuromechanics during balance is also valuable. Balance tasks require sensory response as well as planning and completion of a motor skill (O'Connor & Kuo, 2009). In an MRI study, researchers found that grey and white matter volume increases during a full-body balance task in the frontal and parietal regions of the brain, which was also correlated with balance performance (Taubert, 2010). This provides evidence for brain mechanisms and adaptations that are involved in plasticity and learning during a motor task. Mobile EEG technology has allowed for more development for understanding the connectivity trends during functional movement (Wittenberg et al., 2017). There has been research to show that there are increases in Theta band power during increasingly difficult balance tasks in the frontal and parietal regions (Hülsdünker et al. 2015). These authors suggest that their results emphasize the roles of frontal and parietal cortices in postural control. Further understanding of fundamental organization during a motor task will lead toward empirical evidence for improving movement processing and motor learning. Efficient movement processing is influential in fields like rehabilitation because instructional cues and attentional focus can be applied to enable individuals to be as functionally autonomous as they can.

The current study will examine the superior, central area of the head, which is the area where the primary motor cortex is in the brain, to determine if the brain is activating differently among EFA and IFA, explaining performance differences during EFA. Previous studies using fMRI saw increased activation in the motor cortex when EFA was utilized, in comparison to IFA

(Raisbeck et al., 2020; Wiseman et al., 2020). Additional analysis using EEG can further investigate the underlying mechanisms of attentional focus.

Postural Stability and Motor Performance

Postural stability is integral to maintaining functional independence. In 2018, the CDC reported that 25% of adults ages 65 and older reported a fall (CDC, 2020). The CDC reported that the costs of falls in older adults resulted in $50 billion in medical bills in the United States (CDC, 2021). In North Carolina alone, the costs in 2014 were over $1.2 billion (Florence et al., 2018). These medical costs can include hospital stays, rehabilitation, medical equipment, prescription medication and many more things. Falls not only affects financial situations but overall quality of life. When older adults have a fall, it can lead to decreased mobility, which then has the possibility of leading to negative psychological consequences and/or decreased confidence (Tenetti & Williams, 1998). These outcomes can lead to fear of falling, which will decrease the faller's physical activity and social interactions, leading to even more complications in daily living and quality of life. Activities of daily living, or ADLs, are daily tasks like getting dressed or taking a shower/bath and are used to describe the measure of disability in older adults (Katz, 1983, as cited in Fong & Feng, 2018). The level at which a person can perform ADLs is one way to describe their level of functional independence.

We understand that static and dynamic balance ability declines as we age, especially with disease and/or inactivity (Phelan et al., 2004; Fong & Feng, 2018). Even healthy older adults, postural sway still decreases (Overstall et al., 1977). Therefore, it is important to maintain good postural control for functional independence. Phelan et al. (2004) found that the most common chronic conditions in those with disabilities in activity of daily living (ADL) were Hypertension, Arthritis, Depression, and Diabetes mellitus (type 2) (Phelan et al., 2004). For those with

Diabetes mellitus, peripheral neuropathy is one complication that can occur. Diabetic peripheral neuropathy was found to decrease proprioception, which leads to decreased balance and mobility (Lim et al., 2014). Falls in older populations can lead to other complications like hip fractures. Schnell et al. (2010) studied over 700 older adults with hip fractures resulting from falls and found that those who are over the age of 60, had a 21% overall mortality rate within 1-year of their fall (Schnell et al., 2010). It was then seen in a review by Dyer et al. (2016) that only 40-60% of those who survive a hip fracture after a fall, will regain their pre-fracture mobility level (Dyer et al., 2016). As we age, most experience decreasing mobility, which also lessens physical activity and functional independence. With the increasing rates of falls, there is an increasing need for development of intervention designs and strategies for maintaining and improving function independence.

Most current literature on fall prevention focuses on exercise interventions as well as biomechanics (Espejo-Antúnez et al.2020; Clemson et al., 2012; Cancela et al., 2015). This is important in developing and implementing exercise programming to reduce fall-risk in older adults. To help maintain functional independence the American College of Sports Medicine (ACSM) suggests that older adults should engage in at least 150 minutes of moderate activity every week, that includes the active use of neuromotor skills, such as balance tasks. Balance intervention programs are often used by physical therapists and other practitioners to assist with postural stability and deficits associated with aging. Espejo-Antúnez et al. (2020) and others found that integrating balance focused exercise programs with physical therapy significantly improved functional mobility in older adults (Espejo-Antúnez et al., 2020; Barnett et al., 2003; Clemson et al., 2012; Cancela et al., 2015). Espejo-Antúnez et al. (2020) used proprioceptive exercises in addition to a controlled physical therapy program for 12 weeks in adults who were over the age of 65. They found that the proprioceptive exercise program improved functional mobility, dynamic and static balance, and even reduced fall risk. A study by Barnett et al. (2003) used a 6-month exercise program in adults over the age of 65. The exercises used in this program were very functional, they included sit-to-stand, weight transfer, and balance and coordination types of exercises. The older adults performed the training in a class setting as well as at home. The researchers collected data for 12-month post-intervention on fall risk. They found 40% fewer falls in the exercising group. This study shows how regular exercise for older adults is extremely important for reducing falls and injuries. The benefits of regular exercise in older adults are widely documented for health and wellness related to aging. One study used a combination of strength conditioning as well as flexibility training in older adults (40-70 years old) over one year (Kronhed & Möller, 1998). The researchers found improvements in static balance as well as bone mineral density following the 1-year exercise program (Kronhed & Möller, 1998). There have also been effective exercise programs that focus more directly on balance training. For example, a study by Roger et al. (2003) did a 12-week balance training program where older adults used foam pads and elastic bands to balance and strength train. There were statistically significant improvements in stability in the exercising group, and no changes were seen in the control group (Rogers et al., 2003). There are many different exercise interventions that have shown promising results for fall prevention in older adults. And most of them have potential to be highly adoptable in a variety of settings, this is an important aspect of maintenance of exercise. But what aspects of balance training are more effective than others? It is possible that that there are more effective measures that could be used to improve motor control, without even changing the exercises being used.

The health care system is always looking for cost-effective ways to encourage and integrate effective exercise, and tasks that will lead to maintenance of balance (Smith et al., 2013). Over the past two decades, the literature on attentional focus has clearly identified and supported that when we use an external focus of attention when performing tasks, that motor performance and learning are superior compared to when a different strategy is used (Richer et al., 2017; Ellmers et al., 2016; McNevin et al., 2013; Wulf et al., 2004). If this information was used, via instructional cues, in rehabilitation settings, there is evidence to show that using EFA has the potential to lead to improved postural stability over time. Redirecting attentional focus using instructional cues would fulfill the demand for a cost-effective balance maintenance training tool.

Objective and Subjective Balance Assessment

The subjects will perform the Balance Error Scoring System (BESS) Test which was developed in 1999 by Reimann et al, and is used to assess postural stability (Reimann et al., 1999). It was validated by Iverson et al (2008). It is fast, simple to administer, and relatively inexpensive method to measure postural stability (Iverson & Koehle, 2013). The BESS test is clinically accessible because the only materials needed are a foam pad and a stopwatch. It is often used in concussion assessment to evaluate postural stability (Guskiewicz, 2001; Register-Mihalik et al, 2007; McCrea et al, 2003). To test inter- and intra-rater reliability, Finnoff et al (2009) and others have done observational studies to run statistics and estimate the reliability of the BESS protocol (Finnoff et al, 2009; Bell et al, 2011; Hunt et al, 2009). The general consensus is that the BESS good reliability on certain stances, but overall score was less reliable. A generalizability study by Broglio et al (2009), found the reliability to be moderate. However, the authors found good clinical generalizability using a multiple assessment technique, for example, a pre- and post-test design (Broglio et al, 2009). For the current study, the BESS protocol will be used in conjunction with the BTrackS balance plate.

The BTrackS (Balance Testing System) balance board is a reliable tool to determine postural sway in healthy and clinical populations (Goble et al, 2016; Goble & Baweja, 2018; Benedict et al., 2017). The BTrackS is being used in research to assess for concussion diagnosis using it's built in Sports Balance Software. Goble et al (2016), testing Division 1 athletes, found the BTrackS plate to be highly sensitive and the best instrument for verification of concussions. Goble and Baweja (2018) assessed over 16,000 individuals to obtain normative data for the BTrackS balance testing. The researchers found dependence between the balance testing results and age and sex (Goble & Baweja, 2018). The BTrackS is user-friendly, portable, and more cost effective than most other force plates (Benedict et al., 2017). Compared to a lab grade force plate, the BTrackS is affordable. It has also been validated compared to lab-grade by O'Connor et al (2016). The researchers found high precision and accuracy in the BTrackS, specifically looking at COP data, compared to a lab-grade force plate (O'Conner et al, 2016). The affordability of the BTrackS system can aid in labs or clinics to objectively measure balance. Other balance testing protocols have been validated on the BTrackS balance plate, including the CTSIB. The CTSIB is a protocol that challenges the sensory contributions to postural control. It was found that Goble et al (2019), that the combination of BTrackS and CTSIB were an effective method of assessing balance for diagnosis as well as progression of interventions. Many have used the BTrackS balance plate to assess reliability throughout different balance tasks, and found high validity, reliability, and also accuracy (Levy et al, 2018; Goble et al 2018; Murray et al 2018). The BTrackS balance plate and the BESS protocol will be used in the current study to assess postural control.

Current Gaps in the Literature

External focus of attention is widely known to be more effective in many aspects of motor learning. Much of this research is contrasting EFA and IFA under numerous types of conditions. Although attentional focus literature is widely known, there is a need to understand the underlying mechanisms during performance-based tasks. Data from EEG analysis will serve to find an explanation of the cortical mechanistic characteristics of attentional focus.

The purpose of this study was to determine how attentional focus is reflected in cortical activation, using EEG, during a balance task, by determining if there were different patterns of cortical activity between internal and external focus of attention. There were three hypotheses for this study:

Hypothesis 1a - Postural sway (BESS Score) would be reduced in the external focus condition compared to the internal focus.

Hypothesis 1b - As the balance task increases in difficulty, there would be an increasing Path length score (BTrackS).

Hypothesis 2: Reduced cognitive effort in EF relative to IF would result in greater Alpha (8-12Hz) and lesser Beta (12-25Hz) PSD over anterior electrodes.

Hypothesis 3: Reduced postural sway would be related to greater Alpha (8-12Hz) and lesser Beta (12-25Hz) PSD over medial electrodes.

The independent variable was the mode of attentional focus using instructional cues. The dependent variables were (1) balance performance (BESS Score, BTrackS Path Length) and (2) cortical activity patterns (Alpha and Beta spectral power).

CHAPTER III: OUTLINE OF PROCEDURES

Participants

Thirty-two healthy adults aged 18-35 were recruited to participate in this study from the University of North Carolina Greensboro and surrounding area. Participants were excluded if they had: (1) history of neurological impairments, (2) history of concussion, (3) vestibular or balance disorder. Participants completed an informed consent form and protocol approved by the University of North Carolina at Greensboro's Institutional Review Board

Instrumentation

A foam pad (Airex Balance Pad, Power Systems Inc, Knoxville, TN) were utilized to challenge postural control. BTrackS Balance Plate (Balance Tracking Systems Inc., CA, USA) was being used to measure postural sway.

Task and Procedures

Prior to the start of the study, participants were asked to avoid alcohol, drug use, and caffeine consumption for 4 hours prior to the study. Participants were randomly assigned to an internal focus (n=11), external focus (n=11), or control (n=10) condition group. Figure 1 shows an overview of the study procedures. Participants were fitted with a mobile Electroencephalographic (EEG) system, specifically with 32 electrodes. Electrode placement followed that of the international 10-20 system of electrode placement, with ground and reference. The data was collected using a mobile EEG system at a sampling rate of 500Hz. An impedance test was done (below 50kΩ) ensure a good signal to noise ratio before the data collection took place.

After participants had the EEG cap in place, the amplifier was attached to the participants' posterior trunk. Participants then did a seated, quiet rest period for 3 minutes with eyes open. This quiet rest period is used as a comparative tool.

Postural control was examined using a BTrackS Balance Plate used as a force plate at a sampling rate of 25Hz, totaling 500 data points in each 20-second trial. The path length for postural sway was measured by calculating the total distance travelled by the center of pressure during the twenty-second trial. A greater Path Length suggests decreased postural control, or decreased balance performance.

The balance task performed in this study was the BESS test. The test includes three stances (Double Leg, Single Leg, and Tandem) on both a firm and foam surface, shown in Figure 2. Each stance is maintained for 20 seconds with hands on the hips and eyes closed. The BESS test scores are scored by two examiners from a video recording, by counting the number of errors or deviations from the proper stance using the standardized BESS scoring criteria (Reimann et al., 1999). Testers do not discuss scores with each other. The BESS scores are then compared to Iverson's normative data (Iverson & Koehle, 2013). A higher score is related to decreased performance on the balance test. The firm surface is directly on the BTrackS force plate (BTrackS; Balance Tracking Systems Inc., CA, USA). The foam surface is an Airex Balance Pad (Power Systems Inc, Knoxville, TN) placed on top of the BTrackS force plate. Prior to the start of the experimental protocol, participants were provided with an information session on how to perform the balance task. Following the information session, participants performed all 6 conditions of the BESS test under the assigned experimental conditions of Control, IFA, or EFA. The instructional cues were validated by Diekfuss (2017), during a longitudinal balance intervention study (Diekfuss, 2017). The Control instructional cues will be "Do your best to balance". The EFA instructional cues were "Focus on keeping the foam pad beneath you level and as steady as possible". The IFA instructional cues were "Focus on keeping your feet parallel to the floor and as steady as possible".

Each participant completed the NASA-TLX, used as a measure of perceived cognitive loading in addition to being a quantitative assessment of focus of attention post-intervention and served as a secondary manipulation check (Hart & Staveland, 1988). The NASA-TLX consists of 6 items, and the participant answers on a scale from very low to very high, or from perfect to failure. A higher NASA-TLX score would indicate increased amount of mental effort.

Spectral power activity data was recorded continuously throughout each balance trial using EEG. The path length data from the force plate was used as an objective measure of balance performance. The following compliance check was asked after the last trial: "What were you focusing on during the previous balance tasks?". When the participants report focusing on their body, the answer will be coded as "IFA". When the participants report focusing on the foam surface, the answer will be coded as "EFA". The control group is not given attentional focus instructions. Therefore, to see if there is a natural preference towards one method or the other, they were still asked the manipulation check. The last element the participants completed is the NASA-TLX workload questionnaire.

Figure 1. Schematic of Experimental Procedures.

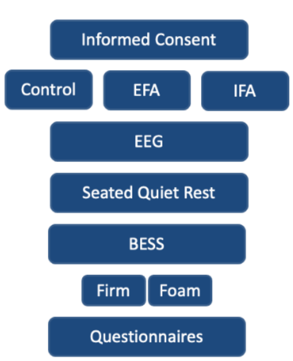
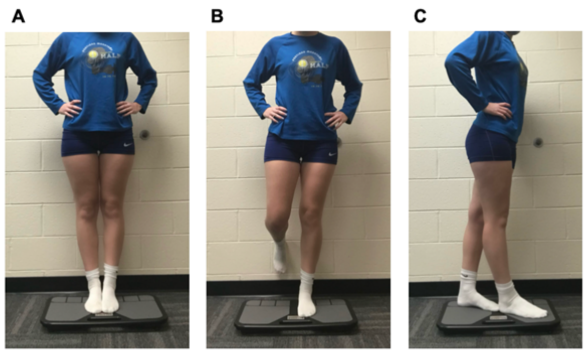


Figure 2. BESS Balance Task.



Note. Firm Surface stances include (A) double leg stance, (B) single-leg stance on non-dominant leg, and (C) tandem stance with non-dominant foot in back. The same stances are used on the foam surface. All stances are done with eyes closed and hands placed on the hips.

Data Collection and Analysis

The mobile EEG system being used was an R-Net electrode system (Brain Products GmpH ©). It is a flexible net with 32 electrodes. The cap is required to be submerged in an electrolyte solution prior to use. The solution consists of KCl, distilled water, and a small amount of baby shampoo. The data acquisition software being used was BrainVision Recorder which wirelessly connects to the amplifier, which is LiveAmp (Brain Products GmpH ©). After data collection was completed, the data was processed using MATLAB (Mathworks Inc., Natick, MA, USA). EEG data preprocessing was done using custom scripts in EEGLAB toolbox on MATLAB (Delorme & Makeig, 2004). EEG is historically done in a static position; therefore, channels were inspected for non-stereotypical movement artifacts when using a mobile EEG system. The data was then analyzed, looking specifically at activation patterns, particularly in the Beta (12-30 Hz) and Alpha (8-12 Hz) frequency bands in the frontal and medial regions. Preliminary electroencephalographic data analysis consisted of cleaning and separating. The preprocessing pipeline code used for this study is in Appendix A. Briefly, the code goes through a preprocessing loop that contains many steps. Early steps included a low and high pass filter, then the first independent component analyses, next the removal of bad channels, followed by a second independent component analysis, next was re-referencing, and finally current source density computation. The PSD was calculated for all the channels for alpha and beta, and then for the 2 ROI's. Electrodes in ROI1 included: 2, 3, 4, 30, 31, 6, 7, 27, 28, 1, 32. They represented the anterior section of the head. Then, ROI2 included these electrodes: 9, 8, 26, 25, 24, 10, 11, 22, 23. They represented the medial section of the head. During the cleaning process some conditions EEG data were found to be insufficient because there were more than two bad ROI channels in either ROI. Bad channels are detected in the code by assigning a fraction detection threshold. This is a fraction of how often a channel must be detected to be rejected in the final rejection. An example of a bad channel would include channels with too much noise. Specific channel information is in Appendix B. The BESS test scores will be used to analyze postural sway during the balance task. Additionally, the BTrackS force plate Path Length was used to measure balance performance. The larger the path length, the worse the performance was. For Hypothesis 1a, a between-subjects ANOVA was used with the BESS score as the dependent variable, and Group (EFA, IFA, Con) as the independent variable. Per Hypothesis 1b, a within-factorial ANOVA was used to assess differences in Path Length compared to balance task challenge. The independent variables are the Group (EFA, IFA, Con) and the Stance (firm: narrow, single-leg, tandem; foam: narrow, single-leg, tandem), the dependent variable is the Path Length score from the BTrackS Force Plate. For Hypothesis 2, linear regression will be used to analyze relationships between Group (EFA, IFA, Con), cognitive effort (NASA-TLX), and power spectral density in the Alpha and Beta bands over anterior electrodes (ROI1). The independent variables are Group (EFA, IFA, Con) and cognitive effort (NASA-TLX) while the dependent variables are Alpha and Beta power spectral density. For Hypothesis 3, linear regression was used to analyze relationships between postural sway (BTrackS Path Length Score) and power spectral density in the Alpha and Beta bands over medial electrodes (ROI2). The independent variable is stance (firm: narrow, single-leg, tandem; foam: narrow, single-leg, tandem) and the dependent variables are PSD (alpha and beta) and Path Length.

.

CHAPTER IV: RESULTS

Primary Analysis

Participants

The 32 participants (24 female; age =22.19 (SD=2.64); mass= 77.82 kg (SD=56.6); height=166.63 cm (SD=11.56)) were of generally healthy condition (no history of vestibular, neurologic, or balance disorder). Characteristics of demographics are displayed in Table 1.

Table 1. Descriptive Statistics of the Sample

Demographics	
Sex	24 female
Average Age (years)	22.19
Average Height (cm)	166.63
Average Mass (kg)	77.82
Regular Smoking	1
Regular Alcohol Consumption	14
Type 1 Diabetes	1
No vestibular/neurological/balance disorders	
No Pregnancies	

Balance

BESS scores had moderate reliability between scorers (IRR=0.594). The BESS data from this study was highly correlated to the normative data (p=0.0001) (Iverson & Koehle, 2013). The data was within a standard deviation of the normative data for this age group (Iverson & Koehle, 2013). Path length scores (BTrackS) increased as the BESS stances increased in difficulty, seen in Table 2. The double-leg firm stance being the easiest, and the single-leg foam stance being the most difficult, according to Path Length scores. The double-leg firm stance has the lowest scores for BESS as well as Path Length. While the single-leg, foam stance has the highest BESS and Path Length scores.

Table 2. Balance Performance: Path Length Score and BESS Score Across Balance Stance Conditions.

BESS Stance	TwoFirm	OneFirm	TandFirm	TwoFoam	OneFoam	TandFoam
Average Path Length Score (cm)	42.59	167.31	117.63	111.28	260.91	209.38
SD	9.74	62.19	52.32	28.35	80.57	87.78
SE	1.72	10.99	9.25	5.01	14.24	15.52
Average BESS Score (# of errors)	0	3.21	0.875	0.13	7	4.75
SD	0	2.72	1.01	0.42	2.29	3.56
SE	0	0.48	0.18	0.07	0.40	0.63

Note. TwoFirm=double leg stance on firm surface. OneFirm= single leg stance on firm surface. TandFirm= tandem stance on firm surface. TwoFoam=double leg stance on foam surface. OneFoam= single leg stance on foam surface. TandFoam= tandem stance on foam surface.

Attentional Focus

Adherence to AF was moderate; for EFA adherence was 73%, and for IFA it was 64% (seen in Table 3). The Reported EF or IF information is based on the feedback given during the manipulation check which takes place after the balance task is complete.

Table 3. Adherence to Attentional Focus Instructions.

	Reported EF	*Reported IF*	*% adherence*
EFA (n=11)	8	3	0.73
IFA (n=11)	4	7	0.64
Con (n=10)	4	6	n/a

Note. EFA=external focus; IFA=internal focus; Con=Control.

Secondary Analysis

Balance and Attentional Focus

There is a moderate effect size on AF and BESS scores (95% CI, partial eta squared=0.056). Shown in Table 4, the IFA group had the highest average BESS score, at 18.45. The EFA and Control groups had very similar average BESS scores, 14.82 and 14.5 respectively.

For most stances there was a small effect size between the Path Length score and AF. The single-foot stance on the firm surface, however, produced a moderate effect size (95% CI, partial eta squared=0.045). Overall, the interaction between Path Length score stance and AF was a low effect size (95% CI, partial eta squared=0.027).

Table 4. BESS and Attentional Focus.

	Average (# of errors)	SD	SE
EFA	14.82	8.05	2.43

IFA	18.45	6.15	1.86
Control	14.5	6.80	2.15

Table 5. Path Length and Attentional Focus.

	Average (cm)	SD	SE
EFA	155.73	34.64	10.44
IFA	151.85	45.93	13.85
Control	146.52	38.34	12.12

Cognitive effort and Attentional Focus

Cognitive effort was determined using the NASA-TLX which indirectly measures overall cognitive workload. Descriptive statistics for the NASA-TLX data are in Table 6. The EFA group resulted in the lowest cognitive effort score, 48.7, compared to the IFA, 54.9, and control, 52.3, groups. Interestingly, the EFA group also had the highest standard deviation, 18.4, meaning the data was most spread out for cognitive effort compared to the other groups.

Table 6. NASA-TLX Cognitive Workload and Attentional Focus.

	Average	SD	SE
Overall Workload	52.0	14.8	2.61
EFA	48.7	18.4	10.4
IFA	54.9	9.46	2.85
Control	52.3	15.9	5.03

EEG

Results of the linear regression results show the relationships between Group, cognitive effort, and power spectral density in the Alpha and Beta bands over anterior electrodes and are shown in Tables 7 and 8. The dependent variables were attentional focus condition (EFA, IFA, Con) and cognitive effort (NASA-TLX) while the independent variables were Alpha and Beta power spectral density in ROI1. Comparing attentional focus condition, power spectral density (PSD) and NASA-TLX scores resulted in no statistical significance (95% CI). The observed non-significance indicates a low chance of a relationship between cognitive effort and attentional focus within the PSD of alpha and beta. In the alpha band, EFA had neither highest nor lowest mean PSD. While in the beta band, EFA had the greatest mean PSD. The IFA group had the highest alpha as well as the highest NASA-TLX score. While the EFA group had the highest beta in ROI1 and the lowest NASA-TLX score.

Table 7. Alpha PSD and NASA-TLX.

	Mean ROI1 (mV) ± SD	Mean NASA-TLX ± SD	R	p
Control	215.85 ± 279.04	52.3 ± 15.9	0.1941	0.6167
IFA	371.27 ± 995.10	54.9 ± 9.46	0.3496	0.322
EFA	309.12 ± 585.11	48.7 ± 18.4	0.1057	0.757

Table 8. Beta PSD and NASA-TLX.

	Mean ROI1 (mV) ± SD	Mean NASA-TLX ± SD	R	p
Control	83.19 ± 89.34	52.3 ± 15.9	0.185	0.6337
IFA	86.37 ± 201.14	54.9 ± 9.46	0.3747	0.2861

EFA	92.39 ± 155.02	48.7 ± 18.4	0.0528	0.8775

Results on the path length and PSD linear regression analysis are summarized in Tables 9 and 10. The independent variable is stance while the dependent variables are mean PSD and mean path length. A linear regression analysis was done to look at the relationship between stance and path length. It was found that the only stance that had a statistically significant relationship to path length was the Single-Foot/Foam-surface stance (onefoam). The onefoam condition is typically the most difficult, indicated by the highest path length. Interestingly, during the onefoam condition both the alpha and beta PSD mean was the lowest of all the conditions. This could imply a relationship between greatest movement and lowered PSD.

Table 9. Alpha PSD and Path Length.

	Mean ROI2 (mV) ± SD	Path Length Mean (cm) ± SD	p	R
twofirm	257.72 ± 659.34	42.75 ± 10.50	0.7519	0.06809
onefirm	651.77 ± 2359.19	166.82 ± 65.41	0.6937	0.07786
tandfirm	466.33 ± 1407.94	114.88 ± 51.70	0.2997	0.2115
twofoam	538.88 ± 1706.87	107.81 ± 24.4	0.6576	-0.09123
onefoam	178.08 ± 181.06	257.57 ± 84.89	0.001856***	0.6133
tandfoam	11976.35 ± 50445.51	65.38 ± 85.07	0.2121	0.2586

***Statistically significant.

Table 10. Beta PSD and Path Length.

	Mean ROI2 (mV) ± SD	Path Length Mean (cm) ± SD	p	R
twofirm	111.12 ± 284.34	42.75 ± 10.50	0.9405	-0.01609

onefirm	225.92 ± 695.41	166.82 ± 65.41	0.7286	0.06862
tandfirm	187.60 ± 562.67	114.88 ± 51.70	0.3334	0.1975
twofoam	181.16 ± 521.90	107.81 ± 24.4	0.6175	-0.1027
onefoam	80.00 ± 63.78	257.57 ± 84.89	0.001383***	0.6264
tandfoam	2240.57 ± 8234.97	65.38 ± 85.07	0.1991	0.2658

***Statistically significant.

CHAPTER V: DISCUSSION

The present study investigated the relationship between attentional focus (EFA, IFA, and Control) and balance (BESS and BTrackS) reflected in possible differences in cortical activity responses (PSD of Alpha and Beta bands; Frontal and Medial regions). These relationships are lightly evaluated previously.

The first hypothesis proposed that the BESS score would be reduced in the external focus condition compared to the internal focus condition. The first hypothesis also proposed that as the balance task increases in difficulty, there would also be an increase in BTrackS score. Regarding the first hypothesis, it was found that postural sway, with respect to the BESS scores, was reduced in the EFA group compared to the IFA. This is confirmed often in other attentional focus studies, EFA is found to be more effective when performing motor skills like balance tasks (Diekfuss, 2017; Kim et al., 2017; Pinto et al., 2021; Park et al., 2015; Richer et al., 2017; Wulf et al., 2004). However, it was found that EFA did not perform better than the control group. It was also found that the path length score increased as the balance task difficulty increased, which indicated objectively poorer performance, with increasing balance demand. The second hypothesis proposed that there would be lower cognitive effort in EFA relative to IFA, that would result in greater Alpha, and lesser Beta PSD over anterior electrodes (ROI1). Cognitive effort is being represented by the scores of the NASA-TLX. Regarding the second hypothesis, the results suggest reduced cognitive effort in EFA relative to IFA. This is a common outcome found in balance and attentional focus research and aligns with the hypothesis (Diekfuss, 2017; T. Kim et al., 2017; Park et al., 2015; Richer et al., 2017; Wulf et al., 2004). In terms of EEG data, the control group had the lowest PSD means in both alpha and beta bands. The EFA group had decreased mean alpha compared to IFA group. This is opposite to the constrained action hypothesis because EFA is thought to insight automaticity which would imply less cortical activity. However, the decreased alpha in the EFA group would relate to increased cortical activity, opposite of the hypothesized patterns. The EFA group had increased mean beta compared to the IFA group. The third hypothesis proposed that reduced postural sway will be related greater Alpha and lesser Beta PSD in across medial electrodes (ROI 2). Regarding the third hypothesis, a general pattern was found over the course of the balances task stances and EEG data. Overall, the highest path length score was aligned with the lowest mean alpha and beta. There was a broad decrease in mean alpha, as the path length score increased. Oppositely, there was a broad increase in mean beta, as the path length path length score decreased. These results are reversed compared to the hypothesis. This may suggest a role of beta activation in medial cortical activity in postural control. In previous research, beta was seen to be greater during a balance task (Baker, 2007; Sanes & Donoghue, 1993). This suggests that the beta band might be involved in processing of movement of the body (Baker, 2007).

The information gained from this analysis has important implications for the design of rehabilitation and motor learning for balance in the future. Firstly, information from EEG data in this study, it can lead to more studies in the future using other brain imaging techniques. It is important to look different points of view from other brain imaging techniques because it would give more full descriptions of what happens in the brain during specific time points. This can aid in the future of rehabilitation by having a more complete idea on why certain techniques may be more effective than others. Secondly, attentional focus is a growing method that can easily be utilized in everyday motor learning. Just a slight change in wording can produce improved performance in balance.

The findings in this study can be compared with the limited number of existing research that examines the relationships between attentional focus, balance, and EEG. One study by Sherman et al (2021) found a weak to moderate relationship between cortical activity differences and balance when looking at Alpha and Theta bands. This suggests an initial relationship between the differences in balance performance and attentional focus with cortical activity. Another study by Ellmers et al (2016) studied the T3-Fz coherence during a balance task, manipulating attentional focus. The authors found a significant effect between attentional focus and this brain region's EEG coherence (Ellmers et al, 2016). Coherence is used to look at interactions between systems in different power bands. The researchers suggested the increase in coherence during the internal focus conditions were due to the use of conscious motor control systems, relying less on automatic control pathways (Ellmers et al, 2016).

This study had several limitations. Specifically, studying EEG signal using an ROI approach comes with many downsides. The main disadvantage to using EEG is its limited, at best, spatial resolution. Although EEG has great temporal resolution, the spatial aspects are low compared to fMRI, for example. The "inverse problem" is the most prevalent issue when deciphering information from EEG data. When studying the brain, it is not rational to assume that one part of the brain is responsible for one specific function. This is because the brain is a complex structure of neurons that cross and collaborate. This is the crux of the "inverse problem". One way the current study addressed this issue was the use of current source density localization. This methodology estimated the current at the scalp which reflects the direction from which the signal might originate. An approach that can be used to avoid using ROI's at all would be network coherence analysis, which determines if two or more electrodes have similar signal activity with one another using algorithmic methods (Bowyer, 2016). Another limitation of this study is the use of the NASA-TLX at the end of the balance task. This questionnaire was given after the hardest balance stance of them all. This may have caused inflated workload scores because participants had just completed the highly demanding of all the tasks. In the future, it might be favorable to also do a workload questionnaire halfway through the balance or motor skill procedure. Another potential solution would be to randomize the balance task order. However, the BESS procedure is used for clinically assessing concussion, so it is not typically used out of its intended order. A limitation of this study was the use of the BESS protocol on top of the BTrackS force plate. The BESS test is validated in a force plate, however, is not validated on the BTrackS force plate, specifically (Riemann et al., 1999). The use of a foam pad on the BTracks is a typical procedure for its built-in programs. This study was not using the BTracks as an assessment of balance, instead it was only used as a force plate. The CTSIB protocol is validated on the BTrackS force plate. This protocol was not utilized in this study because it involves sensory (vision) conditions. This was not ideal for the current study because it was looking at Alpha band activity, which is heavily influenced by whether eyes are open and/or closed.

This study demonstrates the potential for cortical processing differences when focus is manipulated during a balance task. The current study can contribute to the speculation of specific mechanisms of attentional focus benefits to motor skill learning. However, further investigation is warranted to look at more spatial relationships as well as different power bands in temporal EEG data. It is important to break down complex processes like those in this study in order to better understand associative changes and behavioral aspects of motor learning.

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APPENDIX A: EEG PREPROCESSING CODE

```
%% Setup environment
	close all; clear

	SYNKPROJPATH = 'C:\Users\Sydney\Box\SRossback\SYNKEEG';
	SYNKTOOLBOXPATH = sprintf('%s/code/SYNKEEGTOOLBOX',SYNKPROJPATH)
	SOURCEDATAPATH = sprintf('%s/sourcedata',SYNKPROJPATH)
	DERIVATIVEPATH = sprintf('%s/derivatives',SYNKPROJPATH)
	EEGLABPATH = 'C:/Users/Sydney/Box/SRossback/EEG/eeglab/eeglab2022.0'
	load(sprintf('%s/Montage_32.mat',SYNKTOOLBOXPATH))

	if ~exist('ALLEEG','var')
	    cd(EEGLABPATH)
	       eeglab;
	end

	addpath(genpath(SYNKPROJPATH))

	cd(SOURCEDATAPATH)


%% Define necessary variables
	% automatic channel cleaning:
	%   chancorr_crit                         - Correlation threshold. If a channel
	is correlated at less than this value to its robust estimate (based on other
	channels), it is considered abnormal in the given time window. OPTIONAL,
	default = 0.8.
	%   chan_max_broken_time                  - Maximum time (either in seconds or
	as fraction of the recording) during which a retained channel may be broken.
	Reasonable range: 0.1 (very aggressive) to 0.6 (very lax). OPTIONAL, default =
	0.5.
	%   chan_detect_num_iter                  - Number of iterations the bad channel
	detection should run (default = 10)
	%   chan_detected_fraction_threshold     - Fraction how often a channel has to
	be detected to be rejected in the final rejection (default 0.5)
	%   flatline_crit                         - Maximum duration a channel can be
	flat in seconds (default 'off')
	preproccfg.chancorr_crit = 0.8;%
	preproccfg.chan_max_broken_time = 0.3;%
	preproccfg.chan_detect_num_iter = 10;
	preproccfg.chan_detected_fraction_threshold = 0.5;%
	preproccfg.flatline_crit = 'off';
	preproccfg.line_noise_crit = 'off';
	preproccfg.input_filepath = [SOURCEDATAPATH];
	preproccfg.basic_prepared_filename = 'basic_prepared.set';
	preproccfg.preprocessed_filename = 'preprocessed.set';
	preproccfg.channels_to_remove = [];
	preproccfg.eog_channels  = {};
	preproccfg.ref_channel  = [];
	preproccfg.resample_freq = [];
	preproccfg.zaplineConfig = [];
	preproccfg.rename_channels = [];
```

```
        preproccfg.ref = 'CSD'
        preproccfg.filename_prefix = '';
        preproccfg.blocknames=
{'rest','twofirm','onefirm','tandfirm','twofoam','onefoam','tandfoam'};
        force_recompute = 1;


%% Preprocessing loop
% Step 1: Generate list for loops: this is based on the timing files
        list = dir(sprintf('%s/*txt',SOURCEDATAPATH;
        evttimingtxt = {list.name};
            for s = 12:length(evttimingtxt)
            fname = split(evttimingtxt{s},'_');
            sub = convertCharsToStrings(fname{1});
            evttable = importevttimingtxt(evttimingtxt{s},2);

              STUDY = []; CURRENTSTUDY = 0; ALLEEG = []; CURRENTSET=[]; EEG=[];
EEG_interp_avref = []; EEG_single_subject_final = [];

% Step 2: Load data

            subjectfile = sprintf('%s.vhdr',sub);
            [EEG, com] = pop_loadbv(preproccfg.input_filepath,subjectfile);

%Step 3: Initiate the nested 'block' loop & name output path
               for b = 1:size(evttable,1)/2

               Frames=((evttable.Latency/1000)*500); %convert milliseconds to frames
               Frames= round(Frames,0); % round to nearest integer
               oddnums=1:2:14
               blockbeg = Frames(oddnums(b));
               blockend = Frames(oddnums(b)+1);

                sancheck(b,:) = [blockbeg blockend]; %sanity check
                block = convertCharsToStrings(preproccfg.blocknames{b});

              preproccfg.output_filepath =
sprintf('%s/%s/%s',DERIVATIVEPATH,sub,block);

                if ~exist(preproccfg.output_filepath, 'dir')
                    mkdir(preproccfg.output_filepath)
                end


%Step 4: Trim to block & filter
        EEG_filt = EEG;
                EEG_filt.data = EEG_filt.data([1:32],[blockbeg:blockend]);
                EEG_filt.event = []; EEG_filt.urevent =[]
                EEG_filt.chanlocs = EEG_filt.chanlocs(1:32);
                EEG_filt.nbchan = 32;
                EEG_filt.times = blockend-blockbeg;
                EEG_filt.pnts = blockend-blockbeg;
                EEG_filt.xmax = EEG_filt.pnts/500;
                EEG_filt = pop_eegfiltnew(EEG_filt, 1, [], [], 0, [], 0);
```

```
                EEG_filt = pop_eegfiltnew(EEG_filt, [], 50, [], 0, [], 0);
%Step 5: Run ICA #1

                [ALLEEG,EEG_ica,artcomps1,CURRENTSET] =
processMARA(ALLEEG,EEG_filt,CURRENTSET,[0,1,0,0,1])
                EEG_ica.artcomps1=artcomps1;
                icamara_first.(sub).(preproccfg.blocknames{b})=artcomps1


save(sprintf('%s/%s_ica.mat',preproccfg.output_filepath,sub),'EEG_ica')

% Step 6: Identify & Remove bad channels
                [ALLEEG, EEG_preprocessed, CURRENTSET] =
bemobil_process_all_EEG_preprocessing(sub, preproccfg, ALLEEG, EEG_ica, CURRENTSET,
force_recompute);
                EEG_preproc_cat = eeg_regepochs(EEG_preprocessed);

badchans.(sub).(preproccfg.blocknames{b})=EEG_preprocessed.chans2interp;


save(sprintf('%s/%s_preproc.mat',preproccfg.output_filepath,sub),'EEG_preproc_cat')

%Step 7: ICA #2
             [ALLEEG,EEG_clean,artcomps2,CURRENTSET] =
processMARA(ALLEEG,EEG_preproc_cat,CURRENTSET,[0,1,0,0,1])
             EEG_clean.artcomps2=artcomps2;
             icamara_second.(sub).(preproccfg.blocknames{b})=artcomps2;

%Step 8: Re-reference using two approaches (CSD & Avg Ref)

            [G,H] = GetGH(Montage_32,4);
            data = CSD(single(EEG_clean.data),G,H); % compute CSD for <channels-by-
samples-by-epochs> 3-D data matrix
            CSD_3D_final = double(data);
            EEG_cleanCSD = EEG_clean;
            EEG_cleanCSD.data = CSD_3D_final;

            EEG_cleanAvgRef = EEG_clean;
            EEG_cleanAvgRef.data = reref(EEG_clean.data);


save(sprintf('%s/%s_clean.mat',preproccfg.output_filepath,sub),'EEG_clean')

save(sprintf('%s/%s_cleanCSD.mat',preproccfg.output_filepath,sub),'EEG_cleanCSD')

save(sprintf('%s/%s_cleanAvgRef.mat',preproccfg.output_filepath,sub),'EEG_cleanAvgRef
')

%Step 9: Extract spectral power density for all channels in theta,alpha,beta
       %based on Makoto code
            psdOutDb=[]
       [spectra,freqs] = spectopo(EEG_cleanCSD.data, 0,
EEG_cleanCSD.srate,'freqrange',[1 50],'plot','on');
       spectra=spectra(:,[1:50]);
                    thetaIdx = find(freqs>4 & freqs<8);
```

```
                alphaIdx = find(freqs>8 & freqs<13);
                betaIdx  = find(freqs>13 & freqs<30);

        lowerFreq= [4,8,13];
        higherFreq=[8,13,30];
        bands = {'theta','alpha','beta'}
        for p = 1:3
            loF = lowerFreq(p);
            hiF = higherFreq(p);
            meanPowerMicroV = zeros(EEG_cleanCSD.nbchan,1);
            for channelIdx = 1:EEG_cleanCSD.nbchan
                [psdOutDb(channelIdx,:), freq] =
spectopo(EEG_cleanCSD.data(channelIdx,:), 0, EEG_cleanCSD.srate, 'plot', 'off');
                lowerFreqIdx    = find(freq==loF);
                higherFreqIdx   = find(freq==hiF);
                meanPowerMicroV(channelIdx) = mean(10.^((psdOutDb(channelIdx,
lowerFreqIdx:higherFreqIdx))/10), 2);
            end
            PSD.(bands{p})= meanPowerMicroV;
        end
        save(sprintf('%s/%s_PSD.mat',preproccfg.output_filepath,sub),'PSD')


%Step 10: You need to extract power you want from the PSD struct above.

        %%%ROI1 = 2 3 4 30 31 6 7 27 28 1 32 %%
        %%%ROI2 = 9 8 26 25 24 10 11 22 23

        alphaROI1=PSD.alpha([2 3 4 30 31 6 7 27 28 1 32])
        alphaROI2=PSD.alpha([9 8 26 25 24 10 11 22 23])
        betaROI1=PSD.beta([2 3 4 30 31 6 7 27 28 1 32])
        betaROI2=PSD.beta([9 8 26 25 24 10 11 22 23])

        save(sprintf('%s/%s_alphaROI1.mat',preproccfg.output_filepath,sub),
'alphaROI1')
        save(sprintf('%s/%s_alphaROI2.mat',preproccfg.output_filepath,sub),
'alphaROI2')
        save(sprintf('%s/%s_betaROI1.mat',preproccfg.output_filepath,sub), 'betaROI1')
        save(sprintf('%s/%s_betaROI2.mat',preproccfg.output_filepath,sub), 'betaROI2')

        id((7*(s-1))+b,:)={fname{1}}
        blockname((7*(s-1))+b,:)={preproccfg.blocknames{b}}
        meanalphaROI1((7*(s-1))+b,:)=mean(alphaROI1,1,"omitnan")
        meanalphaROI2((7*(s-1))+b,:)=mean(alphaROI2,1,"omitnan")
        meanbetaROI1((7*(s-1))+b,:)=mean(betaROI1,1,"omitnan")
        meanbetaROI2((7*(s-1))+b,:)=mean(betaROI2,1,"omitnan")

               end
            end

        spss=table(id,blockname,meanalphaROI1,meanalphaROI2,meanbetaROI1,meanbetaROI2)
```

APPENDIX B: BAD CHANNEL DATA

Red text are electrodes included in ROI1. Blue text are electrodes included in ROI2.

Participant ID	Block Name	Bad Channels
P01'	rest	[5;29]
	twofirm'	[1;9;19;32]
	onefirm'	[5;6;9;19;29]
	tandfirm'	[1;9;15;19;32]
	twofoam'	[9;19;29;32]
	onefoam'	[1;9;19;32]
	tandfoam'	[5;9;15;19;24;29]
P03'	rest	[1;32]
	twofirm'	[1;3;7;8;11;16;20;24;29;30]
	onefirm'	[1;16;19;20;30;32]
	tandfirm'	[1;16;19;20;30;32]
	twofoam'	[32]
	onefoam'	[11;29;30;32]
	tandfoam'	[3;9;13;16;29]
P04'	rest	[1;24;32]
	twofirm'	[29]
	onefirm'	[3;19;24;28;32]
	tandfirm'	[6;9;24;29;31]
	twofoam'	[3;15;24;29]
	onefoam'	[3;24;28;29;32]
	tandfoam'	[3;5;12;24]
P04'	rest	[24]
	twofirm'	[1;8;12;17;18;20;21;22;23;25;31;32]
	onefirm'	[1;2;12;15;17;24;28;32]
	tandfirm'	[2;3;8;10;18;24;26;29]
	twofoam'	[10;15;24;28;29]
	onefoam'	[8;15;20;24;25]
	tandfoam'	[1;2;12;18;24;29;31]

P07'	rest	[15;19;29]
	twofirm'	[5;15;19;29]
	onefirm'	[15;19]
	tandfirm'	[15;19;29]
	twofoam'	[15;19]
	onefoam'	[9;15;19;32]
	tandfoam'	[9;15;19;29;32]
P08'	rest	[1;9;29;32]
	twofirm'	[1;15;32]
	onefirm'	[1;9;24;28;30;32]
	tandfirm'	[1;6]
	twofoam'	[5;15;29]
	onefoam'	[14;15;18;19;29;32]
	tandfoam'	[1;4;5;9;12;15;24;29;30]
P09'	rest	[1;4;9;24;29;30;32]
	twofirm'	[1;9;24;30]
	onefirm'	[1;5;9;23;29;30;32]
	tandfirm'	[1;5;9;14;19;20;23;27;28;30;32]
	twofoam'	[5;6;9;23;32]
	onefoam'	[1;4;5;8;9;10;12;14;15;16;19;23;28;30;32]
	tandfoam'	[1;5;6;8;9;10;12;15;19;20;23;24;27;28;30;32]
P10'	rest	[5;29]
	twofirm'	[1;9;19;32]
	onefirm'	[1;9;19;24;32]
	tandfirm'	[1;9;19;32]
	twofoam'	[1;9;19;32]
	onefoam'	[1;9;19;32]
	tandfoam'	[1;9;19;24;32]
P11'	rest	[18]
	twofirm'	[15;18]
	onefirm'	[1;18;23;24;29;32]
	tandfirm'	[18;29]
	twofoam'	[18]

	onefoam'	[18]
	tandfoam'	[18;28]
P12'	rest	[1;4;9;24;30;32]
	twofirm'	[1;8;11;16;25;27;28]
	onefirm'	[1;3;6;8;9;16;19;26]
	tandfirm'	[1]
	twofoam'	[6;30]
	onefoam'	[1;6;8;12;15;26;28;32]
	tandfoam'	[6;9;15;19;32]
P13'	rest	[6]
	twofirm'	-
	onefirm'	[9;16;18;19;20;24;28;29]
	tandfirm'	[9;10;24]
	twofoam'	[9]
	onefoam'	[9;23;29;30]
	tandfoam'	[29;30]
P14'	rest	[27]
	twofirm'	[9]
	onefirm'	[9;12;28]
	tandfirm'	[5;9;24]
	twofoam'	[5;9;12]
	onefoam'	[5;9;12;24]
	tandfoam'	[5;9;11;12;24]
P15'	rest	[3;18;19;30]
	twofirm'	[3;5;6;18;28;32]
	onefirm'	[3;15;18;28]
	tandfirm'	[6;15;18;28;31]
	twofoam'	[3;9;10;18;28;31;32]
	onefoam'	[3;14;15;18;24;28]
	tandfoam'	[4;5;9;14;15;18;28]
P16'	rest	[1;9;24;32]
	twofirm'	[1;9;24;28;32]
	onefirm'	[1;4;5;6;9;18;24;28;29;30;32]

	tandfirm'	[1;4;5;9;17;24;28;29;30;32]
	twofoam'	[1;3;5;9;24;28;29;30;32]
	onefoam'	[1;2;3;5;7;9;15;19;24;28;29;32]
	tandfoam'	[1;3;5;6;9;14;18;19;24;28;30;32]
P17'	rest	[9;12;21;22;24;28]
	twofirm'	[2;5;7;12;27;28;30;31]
	onefirm'	[12]
	tandfirm'	[12]
	twofoam'	[12]
	onefoam'	[6;8;12;28]
	tandfoam'	[6;12;28]
P18'	rest	[9;29]
	twofirm'	[6;7;9;22;24;27]
	onefirm'	[7;9;12;24;27;28;32]
	tandfirm'	[1;7;9;19;20;22;24;28;32]
	twofoam'	[4;6;7;9;10;24;27]
	onefoam'	[7;9;19;20;23;24;26;27;28;32]
	tandfoam'	[7;8;12;19;20;24;27]
P19'	rest	[1;6]
	twofirm'	[2;4]
	onefirm'	[4;9]
	tandfirm'	[4;28;30]
	twofoam'	[7;9;23]
	onefoam'	[4;9;27;28]
	tandfoam'	[5;7;9;22;23;25;26;27;28]
P20'	rest	[27;29]
	twofirm'	[1;6;10;24;32]
	onefirm'	[1;5;6;9;24;29;30;31;32]
	tandfirm'	[24;27;32]
	twofoam'	[1;6;24;27;32]
	onefoam'	[5;6;9;14;31;32]
	tandfoam'	[2;3;5;6;16;17;21;24;26;29;31]
P21'	rest	[10;13;14;16;21]

	twofirm'	[9;10;12;13;14;15;19;21]
	onefirm'	[10;13;14;15;19;21;30]
	tandfirm'	[1;10;13;14;15;19;21;29]
	twofoam'	[5;10;13;14;15;19;21]
	onefoam'	[10;12;13;14;15;19;21;28;29]
	tandfoam'	[5;10;11;12;13;14;15;19;21;29]
P22'	rest	-
	twofirm'	[9]
	onefirm'	[3;9;11;32]
	tandfirm'	[1;3;29]
	twofoam'	[1;3;9;13;32]
	onefoam'	[1;12;13;15;22;24;26;32]
	tandfoam'	[1;3;8;12;13;24;27;28;32]
P23'	rest	[5;18;28;29]
	twofirm'	[18;21;32]
	onefirm'	[1;5;6;7;11;18;19;21;24;28;29;30;31]
	tandfirm'	[4;6;18;28;29]
	Twofoam	[2;6;7;9;11;18;21;24;25;28;29]
	onefoam'	[1;3;4;6;9;11;17;18;20;21;28;29;31;32]
	tandfoam'	[1;18;28;29;30;32]
P24'	rest	[32]
	twofirm'	[15;19]
	onefirm'	[15;22;24;29]
	tandfirm'	[15;19;22;29]
	twofoam'	[14;15;19;22;24]
	onefoam'	[1;3;7;9;15;19;20;22;24;32]
	tandfoam'	[1;4;15;19;20;22;32]
P25'	rest	[7]
	twofirm'	[1;7;9;24]
	onefirm'	[7;9;24]
	tandfirm'	[3;5;7;9;23;24;25;28;30]
	twofoam'	[1;3;7;8;9;24;25;29;32]
	onefoam'	[1;6;7;9;24;27;32]

	Tandfoam	[1;2;3;7;9;12;18;20;24;30;32]
P26'	rest	[4;24;28;30]
	twofirm'	[4;9;30]
	onefirm'	[5;9;29;31]
	tandfirm'	[9]
	twofoam'	[1;9;14;24]
	onefoam'	[4;9;29]
	tandfoam'	[4;8;9;19;24]
P27'	rest	-
	twofirm'	[15;20;24]
	onefirm'	[1;9;15]
	tandfirm'	[9;15]
	twofoam'	[9;15]
	onefoam'	[9;15;24;29;32]
	tandfoam'	[5;9;23;29;32]
P28'	rest	[1;9;24;26;28;29;31;32]
	twofirm'	[1;8;9;24]
	onefirm'	[1;2;3;5;9;10;12;24]
	tandfirm'	[1;3;4;5;9;10;12;13;16;23;24;26;28]
	twofoam'	[1;3;5;8;10;12;16;24;28;32]
	onefoam'	[1;2;4;5;7;8;10;12;16;19;23;24;28;30;31;32]
	tandfoam'	[1;2;3;4;13;23;24;28;29;30;32]
P29'	rest	[1;5;6;9;14;15;16;18;20;21;32]
	twofirm'	[6;9;10;16;17;18;21]
	onefirm'	[5;6;10;11;17;18]
	tandfirm'	[6;10;11;16;17;18]
	twofoam'	[6;10;11;17;18]
	onefoam'	[6;8;9;10;11;12;13;16;17;18;24;25;31]
	tandfoam'	[11;16;17;18;25;31]
P30'	rest	[13;16;21;22;24;29]
	twofirm'	[10;12;13;21;22;23]
	onefirm'	[13;21;29]
	tandfirm'	[13;21;29]

	twofoam'	[21;29]
	onefoam'	[22;24;29]
	tandfoam'	[29]
P31'	rest	[5]
	twofirm'	[1;5;6;25;29]
	onefirm'	[1;4;5;6;9;19;21;28;29]
	tandfirm'	[1;3;5;19;29]
	twofoam'	[1;3;5;19;29]
	onefoam'	[1;5;19;24;28;29]
	tandfoam'	[1;2;5;7;15;24;28;29]
P32'	rest	[9]
	twofirm'	[9]
	onefirm'	[7;9]
	tandfirm'	[9]
	twofoam'	[9]
	onefoam'	[7;9]
	tandfoam'	[7;9;14;23;24;30]