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The Influence of Casual Footwear on Human Balance

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THE INFLUENCE OF CASUAL FOOTWEAR ON HUMAN BALANCE

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ABSTRACT

Falls are the third leading cause of unintentional death in homes and communities in the United States, causing 27,800 fatalities in 2012. The ability to maintain postural control is an essential part of activities of daily living (ADLs). However, recent types of casual footwear may be putting the body's postural control system at a functional disadvantage, predisposing wearers to the risk of a fall. The purpose of this study was to examine the effects three forms of casual footwear (thong style flip-flops (FF), clog style Crocs® (CC), and Vibram® Five-Fingers (MIN)) have on postural control following a one mile walk at a preferred-pace. Eighteen healthy male adults (age: 22.9±2.8 years; height: 179±6.0 cm; mass: 81.3±8.8 kg) with no history of neuro-musculoskeletal disorders participated in this study. Static balance measures were recorded using eyes open (EO), eyes closed (EC), eyes open with sway referenced vision (EOSRV), and support, (EOSRP) conditions of the Sensory Organization Test (SOT). The average velocity (VEL) and root-mean-square (RMS) of the center of pressure (CoP) was used to quantify postural sway in the anterior-posterior (APVEL & APRMS) and medial-lateral (MLVEL & MLRMS) directions. Dynamic balance measures were recorded using the medium and large translations of the Motor Control Test (MCT). Muscle activity was collected at 1,500 Hz on the medial gastrocnemius (PF) and tibialis anterior (DF) using a Noraxon EMG system. Mean sway and EMG variables were analyzed using a 3x2 (footwear x time) repeated measures ANOVA. Footwear main effect significance was observed for APRMS in the EC condition (F2,34) = 7.914, p = 0.002), MLVEL (F(2,34) = 3.681, p = 0.036), and APVEL (F(2,34) = 7.469, p = 0.002) in the EOSRV condition with pairwise comparisons displaying increased instability in

the CC and FF compared to MIN. Time main effect significance was seen for MLRMS in the EOSRV ($F(1,17) = 6.532$, $p = 0.02$) and EOSRP ($F(1,17) = 8.982$, $p = 0.008$) conditions with increased instability following the one-mile walk. Future research should seek to elucidate which combinations of these casual footwear characteristics are most detrimental to human balance.

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CHAPTER 1

INTRODUCTION

Maintenance of human balance is an obvious, yet crucial component in order to accomplish simple activities of daily living. The complex task of balance maintenance is exacerbated by the fact that the center of mass (CoM) in human beings is located at approximately two thirds of our body height (Winter, Patla, & Frank, 1990). This design becomes increasingly taxing on our postural control system due to a constantly changing base of support (BOS) such as two feet in contact with the ground (standing/walking), one foot in contact (walking), or no feet in contact (running) in which our center of gravity (CoG) must be kept in order to maintain a balanced state (Winter, 1995). Human balance is sustained by a complex, constant feedback system involving three major sensory systems that work to achieve two primary goals: maintain horizontal gaze with the horizon, and maintain upright vertical body alignment. The postural control system is made up of several subsystems, such as the visual, vestibular, and somatosensory systems. The visual system provides information about the changing environment as well as feedback of our body's orientation as it moves through the environment. The somatosensory system consists of proprioceptors, such as muscle spindles (MS) and golgi tendon organs (GTO), that provide information regarding the orientation of joint segments relative to each other as well as the environment. Also encompassed in the somatosensory system are cutaneous receptors that detect sensations such as touch, temperature, pressure distributions and noxious stimuli. The vestibular system is composed of the structures of the inner ear, its primary goals are detecting linear and angular accelerations of the head and using these detections to maintain the level gaze of the eyes and an upright vertical body alignment in the presence of gravity. The vestibular system also works as a final reference point for making corrective postural adjustments when the visual and somatosensory systems receive conflicting input (Science & Hill, 1996; Winter et al., 1990; Winter, 1995).

The effect of footwear on balance

The foot serves as the direct interface between the human body and the external supporting surface. Any factors that directly influence this foot/floor interface could pose detrimental effects to our body's ability to maintain balance. The design of footwear has been extensively studied as to how it may impact human balance and gait characteristics. Major footwear properties include heel height, heel-collar height, sole hardness, heel and midsole geometry, and slip resistance of the outer sole (Menant, Steele, Menz, Munro, & Lord, 2008; Menz & Lord, 1999). However, in recent years there has been a large influx of casual footwear options for the recreational population, such as the thong style flip-flip, clog style Crocs®, and Vibram® Five-Fingers. The term shod refers to a form of modern footwear commonly characterized by a softer midsole, elevated heel, and typically some form of motion control device built into the shoe (Gangemi, 2011; Shroyer, 2009). Thong style flip-flops have been associated with adverse effects on the lower leg and foot, such as abnormal kinematics and muscle activation measures. However, there is a lack of literature on how these proposed abnormal changes affect standing balance after walking (Shroyer, 2009). The Crocs® are another popular casual footwear choice, often selected because of perceived comfort provided to the wearer. Literature exists to suggest that the textured insoles of Crocs® could provide increased somatosensory feedback and improved balance measures, but there is still a lack of evidence to fully support that notion (Dixon et al., 2012; Hatton, Dixon, Martin, & Rome, 2009). The Vibram® Five-Fingers minimalist shoe has recently become increasingly popular as part of the minimalist movement in footwear design. This design attempts to mimic the barefoot experience while still providing a layer of protection. These footwear should provide just enough protection to the feet to run barefoot without worrying about puncture wounds, cuts and bruises, however there is still a lack of scientific evidence to support these protective claims (Gangemi, 2011;Squadrone & Gallozzi, 2009).

Previous literature has shown the adverse effects footwear can have on postural control measures (Hijmans, Geertzen, Dijkstra, & Postema, 2007; Menant et al., 2008) and while many common forms of footwear have been examined extensively, these recent casual footwear options still lack scientific research to confirm if they are good choices for the health of the population wearing them. Therefore, the purpose of this study is to examine the effects that three popular forms of casual footwear, thong style flip-flops (FF), Crocs® (CC), and Vibram® Five-Fingers minimalist shoes (MIN), have on sway route mean square (RMS) and sway velocity as well as reaction time to an external perturbation following a one mile walk at a preferred walking pace.

Hypotheses

Balance Hypothesis

Specific Aim 1:

To investigate the effects on balance, before and after a one mile walk test, using the sensory organization test (SOT) based on the sensory conflict hypothesis, in which the individuals are challenged with conflicting unreliable visual and proprioceptive sensory information.

H_{01}: There will be no differences in AP/ML sway RMS, and AP/ML sway velocity between pre and post measures of a one mile preferred pace walk.

H_{A1}: There will be significant increases in AP/ML sway RMS, and AP/ML sway velocity between pre and post measures of a one mile preferred pace walk.

The sensory systems of the human body collectively work to maintain a safe and balanced state of equilibrium between the body's segments and the external environment. Previous research has shown that when exposed to heavy workloads, these systems may be impaired (Gribble & Hertel, 2004; Nardone, Tarantola, Giordano & Schieppati, 1997). Detriments to any of these systems can cause hazardous effects to the body's postural control system (Winter, 1995). However, it is not as well known how these compensatory changes will present when the body is exposed to light workloads such as walking, or the effect, if any that the choice of footwear will have (Thomas, VanLunen & Morrison, 2012).

Specific Aim 2:

To investigate the effects on reaction time, before and after a one mile walk test, based on external perturbations using the motor control test (MCT) providing an external perturbation via a moving base of support for analysis of response time of center of pressure adjustments in order to maintain static equilibrium.

H_{02}: There will be no differences in reaction time latencies between pre and post measures of a one mile preferred pace walk.

H_{A2}: There will be significant increases in reaction time latencies between pre and post measures of a one mile preferred pace walk.

For optimal balance, the ability of the postural system to adapt to perturbations from the environment is necessary to maintain stability. This ability of the postural system to respond to lower magnitude perturbations has been shown to be exasperated following physical activity (Gribble & Hertel, 2004; Nardone et al. 1997). However, previous research has induced varying levels of fatigue to the participants in order to observe these effects. It is still uncertain whether constant, low-level perturbations that one would experience in everyday life, such as walking, would cause similar balance decrements without inducing fatigue (Thomas et al., 2012).

Footwear Hypothesis

Specific Aim 3:

To investigate the effects of a preferred pace, one mile walk while wearing casual footwear (FF, CC & MIN) on balance using the sensory organization test (SOT) based on the sensory conflict hypothesis, in which the individuals are challenged with conflicting unreliable visual and proprioceptive sensory information.

H_{03}: There will be no differences between different footwear conditions in AP/ML sway RMS, and AP/ML sway velocity while exposed to a one mile preferred pace walk.

H_{A3}: There will be significant increases among different footwear conditions differences in AP/ML sway RMS, and AP/ML sway velocity while exposed to a one mile preferred pace walk.

The feet act as the direct contact point from the body to the external environment, acting as the base of support for the body and because of their sensory input for center of pressure adjustments, are fundamental in the maintenance of human balance. Footwear, acting as a medium between the foot and the surface, may alter somatosensory feedback from the foot and ankle, as well as masking cutaneous input and pressure distribution from the sole of the foot. Previous literature has shown that footwear properties, such as a secured heel, textured insoles, and midsole hardness can influence characteristics of postural control. The lack of a secured heel does not allow the foot to move as one rigid segment, facilitating insoles may provide increased cutaneous sensory input, and midsole hardness could affect stability by center of mass fluctuations as well as masking kinesthetic feedback of the plantar sole of the foot.

Specific Aim 4:

To investigate the effects of a preferred pace one mile walk while wearing casual footwear (FF, CC & MIN) on reaction time to external perturbations using the motor control test (MCT) providing an external perturbation via a moving base of support in order to analyze response time of center of pressure adjustments in order to maintain static equilibrium.

H_{04}: There will be no differences between footwear conditions in individual's reaction time latencies to external perturbations.

H_{A4}: There will be significant increases between footwear conditions in individual's reaction time latencies to external perturbations.

Footwear is associated directly with the foot-floor interface, and thus, will inherently have some alterations to the body's sense of the external environment through this interface. When the body is tasked with responding to an external perturbation, the sensory systems must work in unison to adequately respond and maintain balance. There is previous literature to support the idea that footwear properties can alter somatosensory feedback, cutaneous input, as well as foot and ankle range of motion. The purpose of this study is to examine the effects these characteristics might have on the body's ability to respond to external perturbations.

Operational Definitions

Posture:

Posture is the relative position of the various parts of the body with respect to one another (the egocentric coordinate system) and to the environment (the exocentric coordinate system). A third frame of reference is that of the gravitational field (the geocentric coordinate system). The orientation of the body part can be described in terms of each of these frameworks (Kandel, Schwartz, & Jessell, 2000).

Postural Equilibrium:

Regulation of posture with respect to gravity is important in maintaining postural equilibrium, which can be defined as the state in which all forces acting on the body are balanced so that the body rests in an intended position (static equilibrium) or is able to progress through an intended movement without losing balance (dynamic equilibrium) (Kandel et al., 2000).

Balance:

The ability to maintain the vertical projection of the center of mass within the base of support. While balance and postural stability are often used synonymously, postural stability depends on the intentional action, the choice of movement strategy and the underlying neuromotor process (Levangie & Norkin, 2011). The maintenance of the center of gravity within the base of support (Winter et al., 1990).

Fatigue:

Muscular fatigue may be defined as an inability of the muscle to maintain a reasonably expected force output (Gribble & Hertel, 2004). A decline in the capacity to generate force

(Corbeil, Blouin, Bégin, Nougier, & Teasdale, 2003). A cognitive perception of tiredness (Cham & Redfern, 2001).

Center of Mass (CoM):

Center of Mass is defined as the point where the three mid-cardinal planes of the body meet, not necessarily located in the body (Rodgers & Cavanagh, 1984).

Center of Gravity (CoG):

Center of Gravity is defined as the point at which the weight force (mg) of a body or system should be applied to a rigid body or system to balance exactly the translational and rotational effects of gravitational forces acting on the components of the body or system. Also known as, the point at which the weight of the body or system can be considered to act (Rodgers & Cavanagh, 1984).

Line of Gravity (LoG):

Line of Gravity is defined as the perpendicular line towards the ground from the center of gravity of that particular body (Levangie & Norkin, 2011).

Base of Support (BoS):

Human being's base of support is defined by the area bounded posteriorly by the tips of the heels and anteriorly by a line jointing the tips of the toes, and is considerably smaller than the quadruped base of support (Levangie & Norkin, 2011).

Center of Pressure (CoP):

Center of Pressure is defined as a quantity, collected by a force platform that describes the centroid of the pressure distribution over a given area (Rodgers & Cavanagh, 1984).

Dynamic Posturography/Sensory Organization Test (NeuroCom):

A testing system which isolates inputs of the vestibular, visual, and somatosensory systems; isolates neuromuscular outputs; and isolates mechanisms of center integration used for postural control and balance (Guskiewicz & Perrin, 1996).

Proprioceptive System:

Sensory system which provides body/limb position and contributes to the maintenance of balance; includes input from the musculoskeletal system (muscles, tendons, and joints); sensory receptors such as muscle spindles and golgi tendon organs that supply information for changes in muscle length and rate of change of muscle length (Kandel et al., 2000).

Visual System:

Sensory system which provides environmental information via the eyes as well as input about movements and position of the body (Kandel et al., 2000; Winter, 1995).

Vestibular System:

Sensory system composed of the structures of the inner ear that detects linear and angular accelerations of the head. Regulates body alignment and head position in the presence of gravity, as well as regulating eye movement (Iurato & Flock, 1967; Kandel et al., 2000).

Clog Style Crocs®:

According to U.S. Patent No. 6,993,858 issued in 2006, clog style Crocs® are a footwear piece comprising the following characteristics. A base section including an upper and a sole formed as a single part manufactured from a moldable foam material, a strap section formed of a moldable material that is attached at opposite ends thereof to the upper of the base section with plastic connectors such that the moldable foam material of the strap section is in direct contact with the moldable material of the base section and pivots relative to the base section at the connectors. Wherein the upper includes an open rear region defined by an upper opening perimeter, and wherein frictional forces developed by the contact between the strap section and the base section at the plastic connectors are sufficient to maintain the strap section in place in an intermediary position after pivoting, whereby the strap section lends support to the Achilles portion of the human foot inserted in the open rear region, and wherein the upper includes a substantially horizontal portion and a substantially vertical portion forming a toe region that generally follow the contour of a human foot, wherein the toe region tapers from an inner area of the base section where the larger toes exist to an outer area of the base section where the smaller toes exist and wherein the sole includes a bottom surface having front and rear tread patters longitudinally connected by a flat section (Seamans, 2006).

Vibram® 5-Fingers Minimalist Shoes:

According to U.S. Patent No. 7,805,860 assigned to Vibram® in 2010, a footwear is provided including the following. A sole and an upper, wherein the sole and the upper delimit the individual toe portions configured to receive, retain, and allow independent articulation of corresponding individual toes of a foot inserted in the footwear, and wherein the sole includes an extension portion which extends upwardly around at least a portion of the foot and the extension portion of the sole comprises a sole toe extension which extends around a front of at least one of said individual toe portions and which extends over at least a portion of a toe nail area of said individual toe portions (Fliri, 2010).

Thong Style Flip Flops:

According to U.S. Patent No. 4,051,610 issued in 1977 a thong style flip flop can be described as a sandal with a sole member and a removable thong member attached to the sole member for holding the sandal wearer's foot against the sole (Shigeji, 1977)

CHAPTER 2:

REVIEW OF LITERATURE:

The purpose of this study is to examine the effects of diverse types of casual footwear on static and dynamic balance. This chapter will provide a basic understanding of postural control, including some of the neuromuscular factors associated with these phenomena. This chapter will be divided into four major parts. First, a review of the prevalence, risks and consequences of falls. The second portion of this review will examine balance parameters and the different bodily systems involved in maintaining balance. Third, is a discussion of varying factors that can influence postural control and balance, such as fatigue, muscle damage or pathological effects. Finally, this chapter will review the previous literature on casual footwear, comparisons to a more traditional form of footwear, and the implications of their growing popularity.

Implications & Prevalence of Falls

Falls are one of the leading causes of unintentional injuries in the United States, accounting for approximately 8.9 million visits to the emergency room in 2009. In homes and communities, falls are the second-leading cause of unintentional death, resulting in more than 25,000 fatalities in 2009 (National Safety Council [NSC], 2011). Falls are common on school campuses, Ballance et al. (1985) showed that falls on the same level not only represented one of the most significant causes of injuries but also were responsible for the largest number of reported accidents at 25 UK universities during 1981-1983. The institutional records for the Ta-Hwa Institute of Technology (THIT; approximate student population of 9500) indicate that there were approximately 300 reported injuries due to slipping and falling during the academic years of 2002-2004. Slipping and falling also accounted for 36-71% of all student injuries on the campus during the same periods (Ballance, Morgan, & Senior, 1985; Li, Hsu, Chang, & Lin, 2007). These injuries can be attributed to lower intensity workloads such as walking (Milroy, Wyrick, Bibeau, Strack, & Davis, 2012).

Balance

Human beings present a very unique and problematic design in terms of balance. The fact that we are bipeds and move with one foot in contact to the ground(walking), no feet in contact (running), or both feet in contact (standing/walking) creates a major challenge to our postural control system (Winter, 1995). The maintenance of balance is important for all animals, but is particularly challenging for humans because of our unique structure. Approximately two thirds of our body mass, including delicate internal organs, are located at two thirds of our height, over two narrow legs which provide a constricted base of support. This design places our postural control system under much higher demand (Winter et al., 1990). Posture can be described as the relative position of the body segments and their relation to each other, the environment and that of the gravitational field. These coordinate systems are termed egocentric, exocentric, and geocentric, respectively (Di Fabio & Emasithi, 1997). Maintenance of balance requires a complex combination of adaptive bodily systems and these systems' ability to adequately respond to the demands of the environment and the action at hand. Three major sensory systems are involved in balance. Vision is the system primarily involved in planning our locomotion and in avoiding obstacles along the way. The vestibular system senses linear and angular accelerations of the head via sensory organs of the inner ear. The somatosensory system is a multitude of sensors that sense the position and velocity of all body segments, their contact with external objects, and the segments position relative to gravity (Winter, 1995).

It should be mentioned that the input from these sensory systems back to the higher command centers in the body is itself, a complex working body. This input system provides feedback on the condition and changing characteristics of the musculoskeletal system and other body tissues, such as the skin. Sensors collect information on such events as stretch in the muscle, heat or pressure on the muscle, tension in the muscle, and pain in the extremity. These sensors send information to the spinal cord, where the information is processed and used by the central nervous system in the adjustment or initiation of motor output to the muscles (Smith, 1976).

To maintain balance, a person must keep their center of mass (CoM) inside the base of support (BOS) (Adlerton, Moritz, & Moe-Nilssen, 2003; Kincl, Bhattacharya, Succop, & Clark, 2002; Winter, 1995). The most important biomechanical constraint on balance is the size and quality of the base of support, typically consisting of the area around the feet. Any characteristics such as size, strength, range of motion, pain or control of the feet will affect balance. In stance, the limits of stability, or the area over which an individual can move their CoM and maintain equilibrium without changing the base of support are shaped like a cone as shown in figure 1 (Horak, 2006). Because of this, equilibrium is not necessarily a position, but a space determined by the size of the support base and the limitations on joint range, muscle strength and sensory information available to detect the limits.

Figure 1. Normal limits of stability. A healthy man leaning his body's center of mass (CoM) (white dot) towards his forward limits of stability, represented as the area of a cone. The projection of the body CoM over the base of foot support is indicated with a white arrow (Horak, 2006).

The CNS has an internal representation of this cone of stability that it uses to determine how to move to maintain equilibrium. Sensory information from visual, somatosensory and vestibular systems must be integrated to interpret complex sensory environments. As subjects change the sensory environment, they need to re-weight their relative dependence on each of the senses (Horak, 2006).

Visual System

The visual system provides information about body position relative to the stationary environment and moving environment, and in conjunction with the vestibular system is a primary balancing processor in movement. The visual system is highly integrated and associated with efficient gait by providing continuous information regarding foot placement, the moving environment and identifying potential hazards. Typically, people rely on visual and somatosensory inputs when maintaining balance under normal conditions (Guskiewicz & Perrin, 1996). When sudden changes or perturbations are induced, causing a person to change his or her direction of movement or head position, the automatic control mechanism provided by the vestibular input becomes crucial for stabilizing the direction of gaze and ultimately equilibrium (Guskiewicz & Perrin, 1996; Winter et al., 1990). The visual system is typically the first reference point used in making postural adjustments whether it is avoiding hazardous environmental factors or regulating body segment positions (Guskiewicz & Perrin, 1996; Hijmans et al., 2007; Horak, Henry, & Shumway-Cook, 1997; Winter et al., 1990)

Somatosensory System

The somatosensory system receives input from articular, cutaneous, and musculotendinous receptors, as well as sensory receptors known as proprioceptors that send afferent signals regarding changes in muscle length and tension (Gribble, 2004). Proprioceptors are sensory receptors in the musculoskeletal system that transform mechanical distortion in the muscle or joint, such as any change in joint position, muscle length, or muscle tension, into nerve impulses that enter the spinal cord and stimulate a motor response (Smith, 1976). The muscle spindle is an encapsulated sensory receptor found in higher abundance in the belly of the muscle lying parallel to the muscle fibers and connecting into the fascicles via connective tissue. The fibers of the muscle spindle are termed intrafusal compared to the muscle fibers that are termed extrafusal. Their main function is to signal changes in the length of the muscle within which they reside. Changes in the length of muscles are closely associated with changes in the angles of the joints that muscles cross. Thus, muscle spindles can be used by the central nervous system to sense relative positions of the body segments. Each spindle has three main components: 1) a group of specialized intrafusal muscle fibers in which the central regions are non-contractile; 2) large-diameter myelinated sensory endings that originate from the central regions of the intrafusal fibers; and 3) small-diameter myelinated motor endings that innervate the polar contractile regions of the intrafusal fibers. When the intrafusal fibers are stretched, the sensory endings are also stretched and increase their firing rate. Because muscle spindles are arranged in parallel with the extrafusal muscle fibers that make up the main body of the muscle, the intrafusal fibers change in length as the whole muscle changes. Thus, when a muscle is stretched, the activity in the sensory endings of muscle spindles is increased. When a muscle shortens, the spindle activity decreases (Smith, 1976). There are equivocal findings in the literature regarding neural activity and varying workloads. Current hypotheses state that following moderate and vigorous workload intensities, muscle fatigue may impair the proprioceptive properties of joints by increasing the threshold of muscle spindle discharge, disrupting afferent feedback, and subsequently altering conscious joint awareness. Other explanations could be that when the hip and/or knee were fatigued, a greater reliance was placed on the ankle to make the small compensatory muscle actions that occur during an ankle strategy. Localized fatigue may reduce efferent signaling, thus reducing the amount of postural correcting muscle contractions. However, it is not as well known whether or not these compensatory effects by the postural control system will be observed after lower intensity workloads such as walking (Gribble & Hertel, 2004; Lundin, Feuerbach, & Grabiner, 1993; Yaggie & McGregor, 2002)

Another important proprioceptor significantly influencing muscular action is the Golgi tendon organ (GTO). This structure monitors force or tension on the muscle. The GTO lies at the musculoskeletal junction. It is a spindle shaped collection of collagen fascicles surrounded by a capsule that continues inside the fascicles to create compartments. The collagen fibers of the GTO are connected directly to extrafusal fibers from the muscles (Smith, 1976). Two sensory neurons exit from a site between the collagen fascicles. When the collagen is compressed through a stretch or contraction of the muscle fibers, the type Ib nerve endings of the GTO generate a sensory impulse proportional to the amount of deformation created in them. Several muscle fibers insert in one GTO, and tension generated in any of the muscles will generate a response in the GTO. In a stretch of the muscle, the tension in the individual GTO is generated along with all other GTOs in the tendon. Consequently, the GTO response is more sensitive in tension than in stretch. This is because the GTO measures load bearing in series with the muscle fibers but is parallel to the tension developed in the passive elements during stretch (Jansen, Rudford, 1964). Damage to any of these systems will affect the overall output of the postural system and thereby a decrement in balance performance (Lepers et al., 1997). Several reports have demonstrated that muscle spindle and Golgi tendon organ activity may be decreased with following vigorous physical activity. Graham et al, in a cat model, demonstrated that large diaphragmatic afferent resting discharge Golgi tendon organs and muscle spindle was reduced under ischemia, electrically induced fatigue, and local acidosis (Graham, Jammes, Delpierre, Grimaud, & Roussos, 1986). Lagier-Tessonnier et al further supported these results by demonstrating that muscle spindle and GTO responses to high frequency vibrations were reduced under conditions of muscle acidosis, ischemia, and hypoxia in the tibialis anterior (Lagier-Tessonnier, Balzamo, & Jammes, 1993). Pedersen et al (1998) demonstrated that fatigue of the medial gastrocnemius resulted in a decrease in the accuracy of information from the muscle spindles in the heteronymous lateral gastrocnemius. Together, these studies support the direct role of fatigue in affecting muscle spindles and Golgi tendon organs and imply a role for fatigue in affecting proprioception following moderate and vigorous physical activities. However, a dearth of literature still exists on the effects, if any, that transient, lighter intensity workloads will have on the body's sensory systems (Graham et al., 1986; Lagier-Tessonnier, Balzamo, & Jammes, 1993; Pedersen, Ljubisavljevic, Bergenheim, & Johansson, 1998).

Vestibular System

Within the inner ear is a vestibular labyrinth which comprises five receptor organs that, complemented by the contralateral ear, can measure linear acceleration along any axis and angular acceleration about any axis. The labyrinth is adjacent to and continuous with the cochlear duct of the inner ear and also consists of three semicircular canals and two large chambers known as the utricle and the saccule. Linear accelerations caused by bodily movements or due to gravity are detected by the utricle and the saccule, while an angular acceleration caused by a rotation of the head or body is detected and measured by the semicircular canals. Information from the vestibular system can be used in three different ways. First, the information is used to control eye musculature in order to keep the eyes fixed on a point as the head changes position. Thus, when the head is suddenly tilted, signals from the semicircular canals cause the eyes to rotate in an equal and opposite direction to the rotation of the head. This is a function of the vestibule-ocular reflex. Second, vestibular information can be used to maintain upright posture, and a third use of vestibular information involves conscious awareness of the body's position and acceleration after information has been relayed to the cortex by the thalamus (Guskiewicz & Perrin, 1996). The vestibular system is the slowest of the afferent sensory systems and is often the last reference point the body will use to make postural adjustments. As well as aiding the visual system with movement information, the vestibular system also works as a reference whenever the visual system and/or somatosensory system receive conflicting input (Iurato & Flock, 1967; Winter, 1995). In short, the vestibular system mainly contributes to the maintenance of balance by maintaining reflexes associated with keeping the head and neck in the vertical position and allowing the vestibule-ocular reflex to control eye movement (Guskiewicz & Perrin, 1996).

Inverted Pendulum Model

The postural system must meet three main challenges. It must maintain a steady stance (balance) in the presence of gravity, it must generate responses that anticipate volitional goal directed movements, and it must be adaptive (Hijmans et al., 2007; Horak, 2006; Winter, 1995). The vertical projection of the center of mass (CoM) onto the ground is called the center of gravity (CoG). Center of pressure (CoP), is the point of location of the vertical ground reaction force vector and represents the weighted average of all the pressures over the surface of the area in contact with the ground and is independent of the CoM.

Postural sway is usually described as a corrective mechanism in response to the external perturbations that are placed on the body. An inverted pendulum model is typically used to describe postural sway, which bears a resemblance to a bilateral quiet stance with the ankle joint acting as the axis of rotation, along the sagittal plane. When the CoG vector is ahead of or anterior to the CoP vector, the body will undergo a clockwise angular velocity and angular acceleration, as seen in figure 2. In order to counteract this forward sway, a plantar flexion moment is performed. This will cause the CoP to move out in front of the CoG, thus causing a counter clockwise angular velocity and acceleration resulting in a posterior movement of the CoG and a backward sway of the body about the ankle joint. This posterior shift in the CoG is sensed by the central nervous system (CNS) and corrective mechanisms are activated, decreasing the planter flexion moment until the CoG lies anterior to the CoP once again. These sequential anterior and posterior moments about the ankle result in what is known as postural sway (Adlerton et al., 2003; Loram & Lakie, 2002; Milton et al., 2009; Winter, 1995).

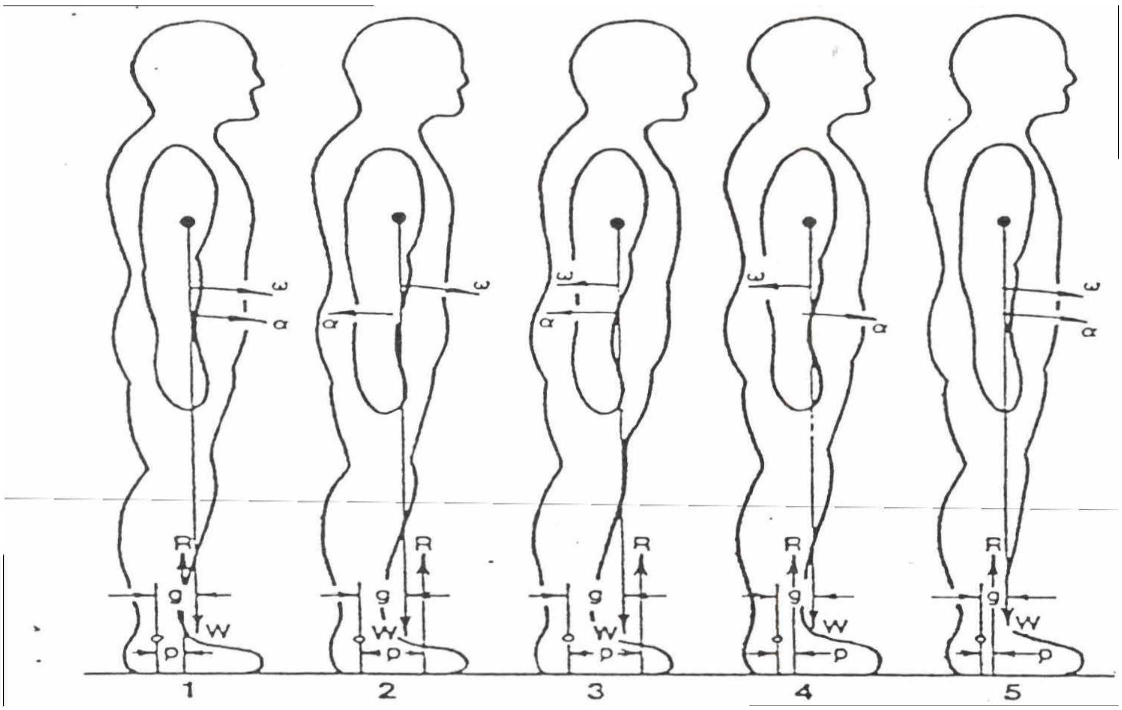


Figure 2. Variations of the CoG versus the CoP during standing showing how the ankle muscles control the CoP and thus continuously regulate the body's CoG (Winter et al., 1990).

The inverted pendulum model can be similarly suggested for postural sway in the medio-lateral (ML) direction, with two ankle, knee and hip joints acting about the frontal plane. The AP control of postural sway is directed by collective effort from the ankle plantar flexors and dorsi flexors about the sagittal plane, while the ML control of postural sway is governed by the collaborative effort of the invertors and evertors. The CoP under each foot will move synchronously back and forth during the AP sway while the CoP during the ML sway will move in the same medial or lateral direction (Adlerton et al., 2003; Horak et al., 1997; Loram & Lakie, 2002; Milton et al., 2009; Winter, 1995).

Three main movement strategies can be used to return the body to equilibrium in a stance position: the ankle strategy, the hip/knee strategy and the stepping strategy. Two of these strategies work while keeping the feet in place and the other changes the base of support through the individual stepping or reaching (Horak, 2006). The ankle strategy, in which the body utilizes more distal musculature first and moves about the ankle in an inverted pendulum model, is the primary method of maintaining balance under small amounts of sway, or when exposed to a small perturbation, and while standing on a relatively firm surface. The hip/knee strategy involves the body using more proximal musculature in order to respond to moderate perturbations. This strategy is used when the ankle strategy alone is not adequate enough in maintaining balance, and works to quickly shift and/or lower the CoM within the base of support. The final strategy in maintaining balance is the stepping strategy. When the other two strategies are not enough to maintain balance an individual will take a step or reach out to touch or grasp an object in order to reestablish the base of support and maintain the CoM within that base of support (Horak, 2006; Winter, 1995)

Balance and Varying Workloads

Previous investigations have assessed the impact of various modalities of exercise on multiple populations, including healthy young adults (Nardone et al., 1997), and older adults at risk of falling (Menz & Lord, 1999), as well as populations with neurological and musculoskeletal disorders (Hijmans et al., 2007; Horak, 2006; Lundin et al., 1993). It is well established that strenuous exercise intensities will affect standing balance measures (Gribble & Hertel, 2004; Nardone et al., 1997), however, these studies have fatigued participants to varying degrees in order to observe these effects. Nardone et al. (1997) examined the effects two different types of exercise (treadmill walking and cycle ergometer pedaling) would have on sway area and sway path. Participants were 13 healthy young adults (six males and seven females, aged 18-39 years). Pre-test balance measures were assessed at the beginning of each session, where patients stood upright and barefoot with feet together on a platform (Kistler force platform, type 9281B). They were to stand as still as possible with their arms by their sides, performing trials alternating between eyes open and eyes closed conditions. Eight of the subjects participated in a treadmill protocol, requiring subjects to begin with a two minute walk at 3 km/hr at 0% grade as a warm up. The speed was maintained constant while the grade was increased to 7% and held for three minutes. Afterwards, the grade was increased to 14% and the speed was increased to 4 km/h. In the remaining 20 minutes, the grade remained constant and the speed was increased from 4 to 5.5 km.h in four steps of 5 minutes each, until the subjects nearly reached the theoretic maximal heart rate (60% of maximal heart rate, calculated by: 220 (males) or 200 (females) – age (years) x 0.6) (Nardone et al., 1997). Eight participants also participated in the cycle ergometer testing; three of these subjects had also participated in the treadmill sessions. This exercise protocol required the subjects to pedal on a cycle ergometer with a frictionally loaded fly-wheel. The cadence of pedaling was held constant at 60 rev/min with the help of a rate-meter. The exercise duration was 25 minutes. Subjects began with one minute of pedaling at a power output of 0 watts. The work-load was then increased to 35 watts and thereafter held constant for three minutes. Next, the workload was progressively increased in order to approximate in the various subjects the changes in perceived exertion and heart rate obtained during the treadmill protocol. Following both experiments, the standing balance test was assessed again (Nardone et al., 1997). Results showed that physical exercise, specifically treadmill exercise, produced various effects on body sway variables that were dependent on the intensity and visual conditions. The average increase in sway reached about 192% of the control values for the sway area and about 132% for sway path with respect to the control values obtained under the eyes closed condition. During the eyes open condition, the increase was smaller for both variables (153% sway area, 118% sway path). However, most detrimental effects seen as a result of fatigue had dissipated within about 15 minutes post exercise, therefore, deeming these effects seriously detrimental to body equilibrium. The authors hypothesized the elevated fatigue found using treadmill exercise over cycling was due to a large percentage of the gait cycle being performed by eccentric muscle actions of the lower limb muscles, particularly the triceps surae, while during cycling the triceps surae mostly undergoes shortening contraction, leading to speculations that peripheral factors might explain the differences in the influence of the two types of exercise on increasing body sway (Nardone et al., 1997).

Corbeil et al. (2002) induced muscular fatigue of ankle plantar-flexors to examine how it would affect upright quiet stance. Eleven healthy male subjects participated in the study (age 20-34 years), and were evaluated for postural sway measures under four conditions (eyes open, eyes closed, with fatigue, and without fatigue). For the fatigue conditions, a block design-training program was used. Muscular fatigue was induced in the ankle plantar-flexors with repeated plantar-flexion of both legs. Subjects were sitting on a standard ankle flexors training device, and were asked to lift a bar upwards by raising the heels. The bar was loaded with weight and placed on the distal portion of the thigh. A maximal load was first determined by adding weight until subjects were able to perform only a single repetition. Then, subjects were instructed to perform 100 repetitions starting at 75% of their maximal workload with a reverse pyramidal technique in which the load is diminished gradually whenever the subjects were unable to perform plantar-flexion. Using a modified Romberg test, CoP range, mean velocity, standard deviation, and maximal instantaneous velocity in the AP and ML directions were calculated.

Fatigue on average, resulted in an increased CoP velocity of .22 and .26 cm/s for the AP and ML axes, respectively. Observable changes in sway velocity, supported previous findings that balance was affected by generalized fatigue. These results were hypothesized to be associated with discrete control of the postural oscillations required to compensate the motor and/or sensory deficiencies induced by peripheral muscular fatigue (Corbeil et al., 2003).

Thomas, Vanlunen, and Morrison (2012) examined the effects of walking at different speeds on standing balance. Their study examined fourteen physically active young adults (7 males, 7 females; age = 24.79 ± 4.23 years) for changes in postural sway before, during, and after walking trials on a treadmill at three different speeds. The speeds were: preferred walking speed (PWS), 120 % PWS, and 140% PWS. Postural sway data were recorded prior to each walking activity (pre-walking) and at five minute intervals during the walking trial and immediately following the final walking period (Thomas, Vanlunen, & Morrison, 2012). CoP excursion, CoP velocity, and total CoP motion were assessed using a Bertec force plate. For all posture assessments, participants immediately stepped off of the treadmill onto a foam pad positioned on the force plate. Participants were then instructed to adopt a comfortable bilateral stance on this surface with their feet hip distance apart. Conditions performed included two sixty second standing trials, one with eyes open (EO), and one with eyes closed (EC). Immediately following each assessment, participants would step back onto the treadmill and continue walking at the same speed. Their results showed an increase in mean Borg scale values from 2.3 ± 0.76, to 3.2 ± 1.1, to 4.8 ± 1.7 in PWS, 120% and 140% respectively. They also saw changes in anterior posterior sway range (mm) in EO and EC balance trials following each gait speed, 36.6 ± 6.5, 40.3 ± 12.4, 42.6 ± 15.7 and 41.9 ± 14.9 for pre-walking, PWS, 120% and 140% respectively for EO, and 55.2 ± 14.6, 59.6 ± 20.5, 61.2 ± 21.7 and 63.0 ± 23.3 for the same respective gait conditions with

EC. The results of this study found that faster walking speeds were more likely to induce changes in postural control, however, some changes do present following lighter workloads as well. Literature suggests that when exposed to moderate or vigorous physical activity the body's sensory systems will undergo compensatory changes and have detriments in function that could be potentially hazardous to the body's ability to maintain equilibrium. There is still a lack of literature, however, on what effects that transient, lower intensity workloads such as walking will have on the body's balance mechanisms.

Casual Footwear

The three forms of footwear being investigated in this study are Vibram® Five-Finger shoes, thong style flip flops, and Crocs®. These footwear were chosen as casual conditions because of their recent influx into the recreational setting as a shoe choice, as well as their varied characteristics and how they associate with the foot and ankle (Hutching & Hons, 2013; Shroyer, 2009). Footwear anthropometric data were collected including heel height, toe height, toe-box width, heel-box width, weight, sole hardness, and midsole hardness. Hardness measures were taken using the Shore A Durometer. Sole and midsole hardness measures were collected 10 times per footwear in order to obtain intraclass correlation coefficients (ICC) to determine reliability of measurements. SPSS was used to calculate ICC using a one-way random effects model, where Midsole ICC = .947 and Sole ICC = .987.

Flip Flops

Flip-flops have always been a popular choice of footwear among recreationally active populations. According to the National Purchase Diary (NPD) Group in Port Washington, a provider of consumer and retail market research information, men's sports sandal sales in 2003 were up five percent from the previous year while overall footwear sales were down six percent (Wilson, 2004). In addition, the Surf Industry Manufactures Association (SIMA) in 2007 reported that one of the top surf industry trends was sandal sales. The SIMA stated that overall footwear sales were down, but sandal sales were up over $300 million, which was an increase of $50 million since 2004 (SIMA, 2007). Although thong flip-flops are a type of sandal and an increased sale of sandals seen by SIMA does not necessarily mean an increase in the thong style, it is of interest to note that men's thong flip-flop sales in department stores had a fourfold increase from 2002 to 2006 as reported by the NPD Group (Dash, 2006). When asked why people prefer flip flops the answer inevitably will come back as comfort. However, there is anecdotal evidence that flip-flops are not conductive to the health of an individual's lower legs and especially their feet (Shroyer, 2009). Common tendencies associated with wearing thong flip-flops have shown that: (1) individuals wear flip-flops beyond the structural limit of the flip-flop (Shroyer, 2009), this is typically seen as the foot bed being worn out and no cushioning properties left in the EVA foam of the flip-flop. (2) Flip-flops are designed and sold with a one size fits all mentality, in that the characteristics of the shoe itself do not have drastic changes from size to size (Shroyer, 2009) and (3) individuals have a different gait while wearing flip-flops versus shoes (Carl & Barrett, 2008; Shroyer, 2009). This observable variability in gait patterns may lead to compensation of unusual stresses that flip-flop wearers do not encounter while wearing a more traditional shoe such as an athletic sneaker (Carl & Barrett, 2008; Shroyer, 2009). The average anthropometric data for the flip flops that will be used in this study are as follows, collected from sizes 7-13. Heel height = 1.4 cm, Toe Height = 1.4 cm, average sole hardness: 71.4, toe-box/heel-box width average: 11.175cm/9cm, weight: 4oz (114.398g) single, and 8oz (226.796g) a pair, and average midsole hardness of 61.3.

Vibram® Five-Fingers

Recently, there has been a significant footwear movement, referred to as the minimalist movement. Minimalist footwear are designed to simulate barefoot conditions as closely as possible, and their primary function as a footwear is to protect against thermal injuries, as well as puncture wounds to the foot (Gangemi, 2011; Squadrone & Gallozzi, 2009) The perceived rationale behind this movement is a back to basics approach that explores the idea that humans were never meant to wear the common walking, running, and dress shoes which flood the market today. This is commonly referred to as shod. The term shod refers to some level of modern footwear that is typically characterized by a softer midsole, elevated heel, and potentially some form of motion control device built into the shoe. The common accepted notion behind minimalist footwear focuses on the research that experienced, habitual barefoot runners will avoid landing on their heel (Lieberman et al., 2010; Squadrone & Gallozzi, 2009). The natural motion during barefoot running is to land with a mid-foot, or even a forefoot strike (Gangemi, 2011). Typically, a heel strike results in a significantly higher vertical force exerted on the body as opposed to a mid-foot or forefoot strike (Gangemi, 2011). The majority of running shoes have been developed to promote the heel strike, and therefore an unnatural running and gait cycle (Gangemi, 2011; Lieberman et al., 2010). A built up heel on a walking or dress shoe also results in a similar problem. A thick heel on footwear will result in increased dorsiflexion while running, which suggests that when wearing shoes, ankle stiffness increases and could be detrimental as a landing strategy while running (Bishop, Fiolkowski, Conrad, Brunt, &

Horodyski, 2006; Gangemi, 2011; Lieberman et al., 2010). New terms have emerged in minimalist literature such as drop (Gangemi, 2011), and are being used to note differences, in millimeters, between the heel and the forefoot. Zero-drop is the term for absolutely no changes from heel to forefoot, as in barefoot conditions. Current literature states that a drop of 6mm or less is considered minimalist. Typical shod conditions have drops of 10mm or more, and women's high heels have drops which often are more easily measured in inches (Edwards, Dixon, Kent, Hodgson, & Whittaker, 2008; Gangemi, 2011). Rose et al. (2011) examined dynamic balance during single leg landings across three footwear conditions. This was assessed with a jump landing protocol while wearing either Vibram® Five-Fingers shoes (V5), Nike Pegasus running shoes (RS) or barefoot (BF). The participants balanced briefly on their dominant leg on a 10 cm high platform positioned 70 cm from the edge of a force plate. They jumped, with eyes open, onto the center of the force plate, landing on the non-dominant leg, typically used for stability. Subjects were asked to stand quietly and motionless on the landing leg for 10 seconds after landing. The subjects rested 10 seconds between jumps, completing 15 total jumps encompassing 5 from each footwear type. Rose and colleagues hypothesized that balance would significantly improve as footwear was reduced due to more accurate, unfiltered cutaneous inputs from the foot. The medial-lateral (MLSI), anterior-posterior (APSI), and vertical components (VSI) along with the total dynamic postural stability index ($DPSI = \sqrt{(APSI^2 + MLSI^2 + VSI^2)}$), were computed for each landing, using a three-second window beginning at the time of landing. Specifically, calculations were the square root of the mean squared deviation of force from the baseline value for three seconds beginning at the time of landing. The results indicated that APSI, MLSI, VSI, and DPSI were all significantly lower in bare feet than in standard running shoes. In addition, MLSI was significantly lower in BF versus V5. The results suggest that static balance was best in bare feet. This is likely due to increasing filtering of sensory input that results from additional material between the foot and the ground. Results from this study indicate that dynamic balance assessed during a single-leg jump landing task is better in bare feet than in standard running shoes. Interestingly however, no significant differences were seen between Vibrams and standard running shoes (Rose et al., 2011). Shinohara and Gribble (2009) investigated the effects of receiving accurate sensory information and its effects on postural stability. Twenty healthy subjects (11 males, 9 females) age 25.5 ± 2.6 years were asked to complete three testing sessions, to measure static postural control. Subjects were tested under three conditions: wearing five-toed socks, wearing regular socks, and wearing no socks. For each condition, static postural control was assessed on a force plate (model 4060NC; Bertec Corp Inc., Columbus, OH) with the subject in a single-limb stance with their hands on their iliac crests, with eyes open (EO) and eyes closed (EC). During the EO trial, subjects were instructed to focus their vision on a large "X" on the wall 3.5 m in front of them and 1.5 m from the floor. The subjects were instructed to keep the non-test limb off the ground in a comfortable position without the limb touching the ground. Standing as still as possible for 15 seconds, if the subjects hopped on the test limb or touched the ground with the non-test limb, the trial was discarded and repeated. COP data were sampled at 50 Hz. For each condition, the COP data were averaged for the three test trials, both for EO and EC. Following collection, Time-to-Boundary (TTB) variables in both the Anteroposterior (TTBAP) and medio-lateral (TTBML) directions were calculated. The TTB dependent variables were the TTB absolute minimum and mean of the TTB minima. The five-toed sock showed significantly lower TTB values than the no sock condition in the TTBML absolute minimum samples during EO trials and the mean of the TTBML minima samples during EC trials. These results indicate that the five-toed condition was associated with decreased static postural stability when compared to the no sock condition. These results are consistent with the hypothesis that filtering or masking of sensory input by footwear can affect postural stability (Shinohara & Gribble, 2009). The average anthropometric measures for the Vibram® Five Finger are as follows, collected from sizes 7-13. Heel height: 0.4cm, Toe height: 0.3cm, Toe-Box/Heel-Box width: 10.63cm/6.21cm, weight: 6oz (170.097g) each or 12oz (340.194g) a pair, average sole hardness from sizes 7-13: 99, average midsole hardness from sizes 7-13: 62.63.

Crocs®

Compared to the other forms of casual footwear already discussed, there is a relative dearth of literature on Crocs®, and how they influence gait, joint kinematics, or postural control. The research associated with Crocs® as a form of footwear focuses primarily on the textured insole Crocs® provide, examining the effect these insoles have on temporal-spatial gait parameters. Dixon et al. (2012) examined two different textured insoles, comparing the previously used insole (group A) to a commercial Crocs insole (group B). Their findings were that after a two week pre and post design, stride length increased between baseline and follow-up in both legs for both insole groups and group B. Other findings were stride velocity and frequency did not change in either group. However, this research was focusing on multiple sclerosis populations, so making inferences to a general population is inappropriate without further research on textured insoles and their effects on gait and balance parameters (Dixon et al., 2012). A similar study by Hatton, Dixon, Martin, and Rome (2009) examined the effects of quiet standing on different surface textures, on postural stability and electromyography of lower limb musculature. Three different textured surfaces were used, T1 (3mm thickness, EVA foam, pyramidal indentations), T2 (3mm thickness, Lunsasoft Mini Non Slip, convex circular patterning, and T3 (completely flat surface texture, EVA foam, 3mm thickness). Twenty-four healthy adults participated in the study age 20-35 years (mean age 27.5 ± 7.9 years). All participants were tested in barefoot, bipedal quiet standing with eyes open. Each test repetition lasted for 30 seconds, with three trials for each texture condition. Anteroposterior and medio-lateral range and standard deviation (mm) were calculated by the force platform. Results showed no statistically significant differences among the three conditions for any of the postural sway variables: (AP SD, $p = .105$), (AP range, $p = .216$), (ML SD, $p = .669$), (ML range $p = .957$). The results of this study show that two textured surfaces, differing only in their pattern of indentation, did not significantly affect AP or ML postural sway in comparison to the normal control condition. The balance tests in this study however, used sheets of texture rather than insoles. Thus, further research is needed to see the exact effects textured insoles could have on postural control (Hatton et al., 2009). Average clog style Crocs® anthropometric data are as follows, collected on sizes 7-13. Heel height: 2cm, Toe height: 1.6cm, Toe-Box/Heel-Box width average from sizes 7-13: 10.49cm/8.81cm, weight: 7oz (198.447g) each or 14oz (396.893g) a pair, Sole hardness: 51.2, Midsole hardness: 68.9.

Heel Height

Footwear is known to have influencing effects on postural control, and the subsequent risk of slips, trips, and falls by altering somatosensory feedback to the foot and ankle and modifying frictional conditions at the shoe-floor interface (Menant et al., 2008). If the heel of a shoe is raised high enough to substantially alter the position of body segments and therefore the total body's CoM position, compensatory mechanisms for postural changes and whole body kinematics will be utilized. Postural, kinematic, or kinetic changes could result in unnatural joint loading patterns having potentially problematic consequences (Snow & Williams, 1994). Brecht, Chang, Price, and Lehmann (1995) examined the relative balance performance of female participants while wearing cowboy boots and regular tennis shoes. During testing, a subject stood on a computer controlled platform that moves in a linear horizontal direction, and can measure a break in contact to one millisecond. The system was set to move 20 cm during the test and it reached a velocity of 40 cm/s. The platform accelerated was varied from 10 cm/s^2 to 150 cm/s^2. The platform accelerated in the forward direction and caused the subject to sway backwards. Their results showed the heel height between the two footwear types was significant (mean shoe height = 1.87 cm, mean boot height = 3.70 cm, $p < .001$). The length between the heel and the metatarsal contact point on the cowboy boots was also significantly decreased from that of the tennis shoe, effectively decreasing the BoS that the CoM could move within leading to potential for an increased loss of balance and/or fall. They also found that people are more likely to lose their balance and fall posteriorly when wearing cowboy boots than when wearing tennis shoes. The anteriorly tapered heel severely decreases the posterior moment arm, and would require less of a sway on the heels before the person loses balance and starts to fall (Brecht, Chang, Price, & Lehmann, 1995).

EMG measures have been used as well to determine if there are varying levels of muscle activation in the quadriceps (Vastus Medialis and Vastus Lateralis) while performing a sit to stand test in increasing heel heights. Results indicated that there was a link between increased heel height and EMG in both the VM and VL. The 1cm heel did not produce significant increases, the VL showed increases in activity at 3cm and 5cm, and the VM showed significant at 5 cm. Mean EMG activity for VM and VL during sit to stand. These variations in musculature may be due to compensatory postural mechanisms having to account for the changes in limb kinematics while wearing heels of different heights (Edwards et al., 2008). A comparison of these results in the literature presents equivocal findings. Lee et al. (2001) measured EMG of VL and observed that the effect of increasing heel height on EMG during gait was not statistically significant. Differences in the relative levels of EMG activity in VM and VL have been linked with abnormal knee mechanics as well, however, even in studies that have observed increased EMG with high heels, the ratio of VM to VL EMG was not significant (Edwards et al., 2008; Lee, Jeong, & Frievalds, 2001).

Snow and Williams (1994) examined eleven subjects for CoM position changes, forefoot loading, lumbar curvature, and pelvic tilt during standing. The shoes were commercially available shoes of three different heel heights. A low heeled shoe (1.91 cm), a medium (3.81 cm), and a high (7.62 cm). The ML by AP dimensions of the heel of the shoes were 3.4 by 4.9 cm (low heel height), 0.9 by 1.2 cm (medium), and 0.7 by 0.8 cm (high). The postural results of this study show no changes in standing while wearing shoes of various heel heights. Equivocal findings among studies has been hypothesized as variations in participants choices of postural control strategies, as well as the wide range of methods used for quantifying postural control (Brecht et al., 1995; Menant et al., 2008; Snow & Williams, 1994). Ko and Lee (2013) examined the displacement of the CoP and changes in the distribution of foot pressure after walking in flat (0.5cm), middle-heeled (4 cm), and high-heeled (9 cm) shoes for 1 hour. Fifteen Healthy women wearing shoes with heels of each height in a random order participated in the study. An FDM-S (zebris Medical GmbH, Germany) force platform was used to measure the plantar foot pressure and displacement of the CoP for all subjects. The results from this study showed that walking in 4 cm heeled shoes did not significantly change the distribution of foot pressure, however, pressure distributions did significantly change after walking in either 0.5 cm or 9 cm shoes. Similarly, the CoP was not significantly displaced after walking in middle-heeled (4 cm) shoes but was significantly displaced after walking in either flat (0.5 cm) or high-heeled (9 cm) shoes. The distributions of foot pressure shifted toward the hind foot in flat (0.5 cm) shoes and shifted to the forefoot after walking in high-heeled (9 cm) shoes. These results suggest that the distribution of pressure during standing moves more towards the forefoot as heel height increases while walking (Ko & Lee, 2013).

Sole Cushioning Properties

The use of foam materials is a very common manufactured tool to increase the perceived comfort of the wearer. Softer midsoles became common in shoes as a means of reducing impact during gait. Midsole hardness is measured using the Shore classification system of material compressibility, ranged from Shore A15 (softest) to Shore A50 (hardest) (Menz & Lord, 1999; Robbins, Waked, Gouw, & McClaran, 1994; Robbins, Waked, & McClaran, 1995). Footwear with the thickest, softest midsole resembles a modern day athletic shoe, and a footwear with the thinnest, hardest midsole resembles a conventional leather walking shoe (Menz & Lord, 1999). Previous studies have been conducted to investigate the effects of sole and midsole thickness and hardness on stability in older and younger individuals. (Robbins & Waked, 1997; Robbins & Gouw, 1991; Robbins, Hanna, & Gouw, 1988). Robbins, Gouw, and McClaran (1992) evaluated the balance ability of 25 older men (mean age = 69 years old), and demonstrated the detrimental effect that soft and thick shoe midsoles have on balance control by assessing the frequency of falls from a walking beam. Subjects were asked to walk at a fixed speed of approximately 0.5 m/s, on a 9 m long and 7.8 cm wide balance beam without observing their feet. The number of times the subjects stepped off the beam was recorded, as well as balance failure frequency, which was defined as the number of steps from the beam after ten trials. Subjects wearing the thinnest, hardest midsole performed the best at the task. A subsequent study by Robbins, Waked, Gouw, and McClaran (1994) involving 17 men (ages 19-50 years, mean age = 32.6) implemented a similar study design to the balance beam testing previously done on an elderly population (Robbins et al., 1992). This study examined the effects of seven footwear conditions: barefoot and six different types of footwear. The midsoles were measured for hardness on a typical Shore scale, and measured either A15, approximated as the softest midsole, A50, corresponding to the hardest midsole, or A33, representing the mean. Along with each midsole hardness, there were two midsole thicknesses. The thinner midsole was 13mm thick at the heel, and 6.5 mm thick under the metatarsal-phalangeal joint. The thicker shoe's midsole was 27 mm at the heel and 16 mm under the metatarsal phalangeal joint. These two respective thicknesses represent the thickest and thinnest currently available for footwear of this construction (Robbins et al., 1994). Walking speed of the subjects was set constant at .5 m/s, completing ten passes down the beam for each testing condition, the distance from the beginning of the trial to the site of balance failure was estimated to the nearest meter and recorded. Results from this study agreed with that of the previous (Robbins et al., 1992). Interestingly there was no significance in balance failure due to the subject's age, however, there was a significant main effect for midsole thickness, and hardness. Post-hoc t-tests also revealed significant effects for midsole thickness, increasing balance failure frequency by 54.3%. Furthermore, changing from the hardest to softest midsoles increased balance failure frequency by 77.1%. Another interesting result to note is that 88.2% of subjects selected the A15, thick midsole shoe, as the most comfortable, and while none of the footwear in this study were deemed uncomfortable by participants, this gives further credibility to the notion that footwear commonly chosen by the masses for perceived comfort also may cause the most instability. It was speculated that these results may be in part due to masked somatosensory feedback from the plantar surface of the foot, and that the expanded foam midsoles may cause the wearer to be unable to properly judge pressure distribution across the foot (Robbins et al., 1994).

CHAPTER 3:

METHODOLOGY

The purpose of this study was to examine the effects of popular forms of casual footwear on postural stability following a one mile walk at a self-selected pace. Analyses conducted focus on how the footwear variations affect human balance and lower limb muscle activity during quiet standing postural control tests.

Participants

Eighteen healthy, recreationally trained males between the ages of 18 and 44 years were recruited with the use of posted recruitment flyers on bulletin boards on the University of Mississippi Campus, specifically in the Turner Center and the Student Union, as well as by class announcement and email to all the Health, Exercise Science and Recreation Management (HESRM) students. Participants were required to fill out two forms; the physical activity questionnaire (PAR-Q) and seven day physical activity recall (7-day PAQ) in order to determine whether they were healthy enough and physically active enough to participate (Sallis et al., 1985). Participant demographic information is located in Table 1

Table 1

Participant Demographics	Mean ± SD
Age (years)	22.9 ± 2.9
Mass (kg)	81.3 ± 8.8
Height (cm)	179.2 ± 6.0
Moderate to Vigorous Physical Activity (MVPA) (min/week)	266.6 ± 102.9

Instrumentation

Quiet standing CoP measures were analyzed using the NeuroCom® Equitest® Balance Master® – Posture Platform (NeuroCom International, Inc. Clackamas, Oregon). The sensory organization test (SOT) uses participant's CoP to quantify postural sway while somatosensory and visual environments are altered systematically. During the SOT the forceplate, visual surround, or both may be "sway referenced" so that they move to follow the participant's anterior-posterior (AP) sway. Specific pairs of tests compare different mechanisms and sensory systems for balance. The SOT consisted of four testing conditions: standing with (1) eyes open (EO) and (2) eyes closed with the platform and visual surround fixed (EC), (3) standing with the platform fixed, eyes open with the visual surround sway referenced (EOSRV), (4) standing on the platform sway referenced with eyes open (EOSRP). The variables were the sway velocity components in the medial-lateral (M/L) and anterior-posterior (A/P) directions, and root mean square (RMS) of CoP displacement in the anterior-posterior (A/P) and medial-lateral (M/L) directions. Sway velocity (cm/s), is a measure of the angular change of the CoP per unit time, where the value is representative of changes in the location of the CoP in the anterior, posterior, medial, and/or lateral directions. Higher values indicate decreased postural stability, as they imply larger angular changes in the location of the CoP. Previous research has identified sway velocity as an appropriate dependent measure for use in determining postural stability (Wade et al., 2004). RMS (cm) denotes a measure for mean body sway of a specific period of time and a comparison to be made between conditions (Raymakers, Samson, & Verhaar, 2005; Davidson et al., 2004).

During the Motor Control Test (MCT), latencies are quantified by the time between translation onset and the initiation of the subject's active response in each leg. The MCT delivers external perturbations via translations of the force plates, using four different algorithms to determine the latency of the response of each leg during translations, and provides a quality factor number to indicate the consistency of agreement of these algorithms for the averaged responses. To determine self-selected walking pace for all participants, a 70ft indoor walking track was used. Following the one mile walk, observed fatigue was defined as a decline in somatosensory feedback measured by a significant increase in co-contraction index (CCI) (Benjuya et al., 2004; Laughton et al., 2003).

$$\text{SWAY VEL} = \left(\frac{1}{t}\right) \sum_{i=0}^{n} |COP_i - COP_{i-1}| \qquad \text{Equation 1}$$

$$\text{SWAY RMS} = \sqrt{\frac{1}{n} \sum_{i=0}^{n} \left(COP_i - COP_{avg}\right)^2} \qquad \text{Equation 2}$$

For equations (1) and (2), N = total number of data samples, COP_{avg} is the overall average COP position for the duration of the trial and T = the total time of the trial (Wade, Garner, Redfern, & Andres, 2013).

Experimental Conditions

Participants were asked to take part in three separate experimental conditions, wearing three different types of footwear that include a thong style flip-flop (FF), crocs with clogs (CC), and Vibram Five-Fingers minimalist shoes (MIN).

Measures of hardness are a widely used procedure and for determining the material properties of treated plastics and rubbers through the indentation of a test piece. The Shore A hardness scale measures the hardness of flexible mold rubbers that range in hardness from very soft and flexible (Shore A 0-40), to medium and somewhat flexible (Shore A 41-70), to hard with almost no flexibility (Shore A 71-100) (Bassi, 1986). The most commonly used size of footwear used in this study was 11, for which the footwear characteristics are listed in Table 2. For footwear mass, * indicates a significant difference between CC and FF, # indicates a significant difference between CC and MIN, and † indicates a significant difference between FF and MIN.

Table 2

Footwear Characteristics

Shoe	CC	FF	MIN
Mass (g)	364.5*#	226.8†	340.2
Sole Thickness (cm)	2.0*#	1.4†	0.4
Sole Hardness	A51.2*#	A71.4†	A99
Heel Hold	"Partial"	None	"Direct"

Experimental Procedures

A repeated measures, counter-balanced design using within-subjects factor was used. All participants visited the Applied Biomechanics Laboratory and Kevser Ermin Applied Physiology Laboratory at the University of Mississippi four times, each separated by a minimum of 24 hours. A description of the experimental procedures for each visit is provided below.

Day 1: The first visit consisted of a familiarization day, where each participant signed the University approved informed consent, and completed the physical activity readiness questionnaire. After paper-work completion, participants were exposed to the testing measures for balance and muscle activity, and anthropometric measurements such as height, weight, resting heart rate and resting blood pressure. Next, pariticipants were asked to walk on a 70ft indoor track at their preferred walking pace to obtain walking speed. Participants were instructed to walk at their normal, leisure pace for all walking trials.

Day 2: Experimental Testing, Part I:

The participants began at the Applied Biomechanics Laboratory where they were prepared for EMG procedures, by shaving, abrading, and cleaning the skin with alcohol swabs at the electrode sites before electrode placement. Following EMG electrode placement, partcipants performed a pre-test for the following measures in Part I. Participants were tested for muscle activity with EMG during a 5 second maximal voluntary contraction (MVC) for the lower extremity muscles of the dominant leg. Leg dominance was determined by self report. This was performed to obtain MVCs for the medial quadriceps, medial hamstrings, tibialis anterior, and medial gastrocnemius. Maximal voluntary contractions for the quadriceps and hamstrings were performed while seated on a padded weight bench with a leg extension attachment. Participants were asked to extend the lower leg as hard as they could into the padded leg extension attachment 3 times, for 5 seconds a trial, to obtain quadriceps MVC. Similarly, participants were asked to flex the lower leg into the pad 3 times, for five seconds a trial in order to obtain the hamstring MVC. Next, to obtain the tibialis anterior MVC the participant had their foot secured to the weight bench and were asked to dorsiflex as hard as they can for three trials at 5 seconds each. Finally, subjects were asked to plantarflex into the ground, standing on their toes as hard as they can for 3 trials of 5 seconds each in order to obtain the medial gastroc MVC. Following this, participants were escorted to the NeuroCom and instructed to stand as still as possible on the NeuroCom for balance assessment using the Sensory Organization Test (SOT) and the Motor Control Test (MCT).

Experimental Testing, Part II:

The participants then moved to the Kevser Ermin Applied Physiology Laboratory and each participant was evaluated by walking one mile on a treadmill at their preferred pace. This speed was determined by evaluating their pace from 6 timed 70ft trials on an indoor track. Participants were timed over the middle 50 feet during each trial and preferred pace was determined as the mean pace traveled over those 6 trials in a manner previously described (Morris et al., 2014). The participants were escorted to the treadmill and instructed to walk for one mile on the treadmill at the previously described preferred pace

Experimental Testing, Part III:

Following the one mile walk, participants were re-tested for muscle activity with EMG during a 5 second MVC for the lower extremity muscles of the dominant leg. Following this, participants were escorted to the NeuroCom® and instructed to stand as still as possible on the NeuroCom® for balance assessment using the SOT and the MCT.

Day 3 & Day 4 followed a very similar experimental protocol, but with different casual footwear: flip-flops, Crocs®, and minimalist shoes as determined on familirization day by the participants order in the counterbalanced design. All testing days were separated by a minimum of 24 hours.

Statistical Analysis: Results were analyzed in SPSS with a predetermined alpha level of 0.05 using a 2 x 3 [2 time measures (pre, post) x three footwear types (FF, CC, MIN)] repeated measures analysis of variance (ANOVA). Analyses were run on each of the dependent sway parameters for each of the SOT conditions as well as the MCT latencies. Mean EMG, percent muscle activation, and ankle co-contraction index were analyzed individually using a 2 x 3 [2 time measures (pre, post) x three footwear types (FF, CC, MIN)] repeated measures analysis of variance (ANOVA). Surface electromyography (EMG) signals were recorded from the right leg musculature: Vastus Medialis (Q), semitendinous hamstring (H), tibialis anterior (TA), and medial gastrocnemius (MG). The surface EMG signals were recorded using silver/silver chloride monopolar surface electrodes. The ground electrode was placed on the tibial plateau. The EMG was recorded using Noraxon® MyoResearch software (Noraxon U.S.A. Inc. Scottsdale, AZ.). Raw EMG data was collected at 1,500 Hz, Band-pass filtered (20-250Hz) and rectified prior to analysis. Pairwise comparisons with a Bonferroni correction were used to identify post-hoc differences if interaction or main effect significance were found. If at any point during analysis there was a violation of Mauchly's test of sphericity, a Greenhouse-Geisser correction was used to determine significance.

CHAPTER 4

RESULTS

Participants:

Eighteen healthy male adults (age: 22.9 ± 2.9 years; height: 179 ± 6.0 cm; mass: 81.3 ± 8.8 kg) with no history of neuro-musculoskeletal disorders completed this study.

Analysis:

Static balance measures were recorded using the four conditions of the Neurocom® Equitest (EO, EC, EOSRV, EOSRP). The average sway velocity and root-mean-square (RMS) of the center of pressure (CoP) were used to quantify the postural sway in the anterior-posterior (APVEL & APRMS) and the medio-lateral (MLVEL & MLRMS) directions.

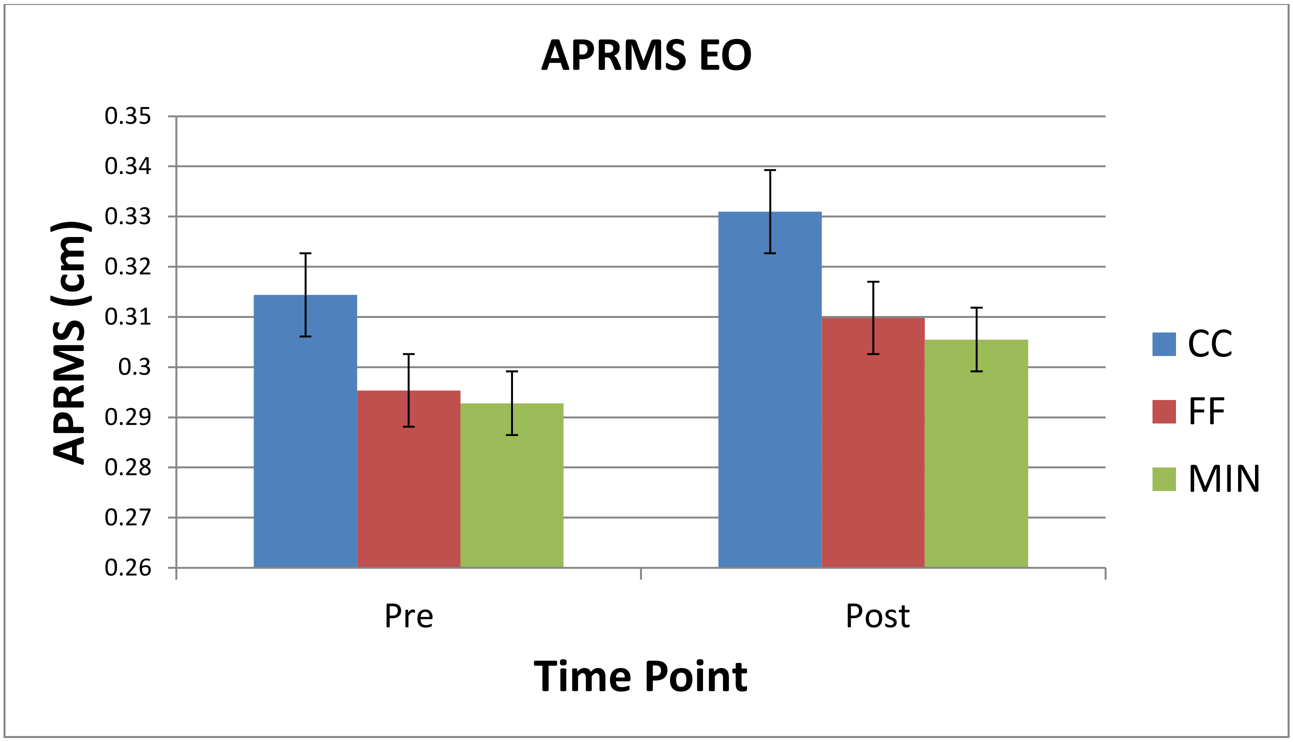
Footwear Characteristics:

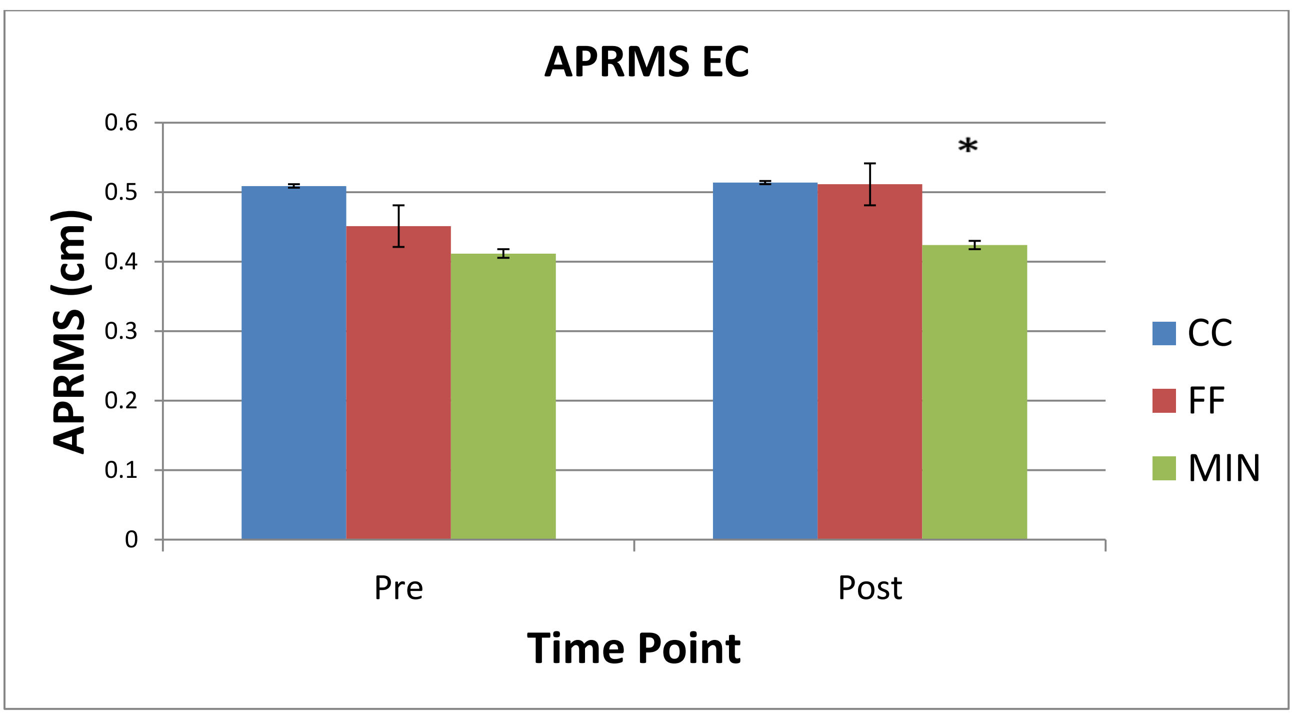
Inidividual independent t-tests were used to analyze the difference between mass, hardness, and sole thickness of the footwear. Significant differences in mass were observed between all pairs of footwear. The Croc® compared to the flip-flop ($t(34) = 36.108$, $p = <0.001$). Croc® compared to the minimalist ($t(34) = 11.877$, $p = <0.001$), and flip-flop compared to minimalist ($t(34) = -25.993$, $p = <0.001$). Significant differences in hardness were observed between all pairs of footwear. The Croc® compared to the flip-flop ($t(34) = -28.348$, $p = <0.001$). Croc® compared to the minimalist ($t(34) = -43.123$, $p = <0.001$), and flip-flop compared to minimalist ($t(34) = -30.016$, $p = <0.001$). Significant differences in thickness were observed between al pairs of footwear. The Croc® compared to the flip-flop ($t(34) = 18.922$, $p = <0.001$). Croc® compared to the minimalist ($t(34) = 51.239$, $p = <0.001$), and flip-flop compared to minimalist ($t(34) = 101.342$, $p = <0.001$).

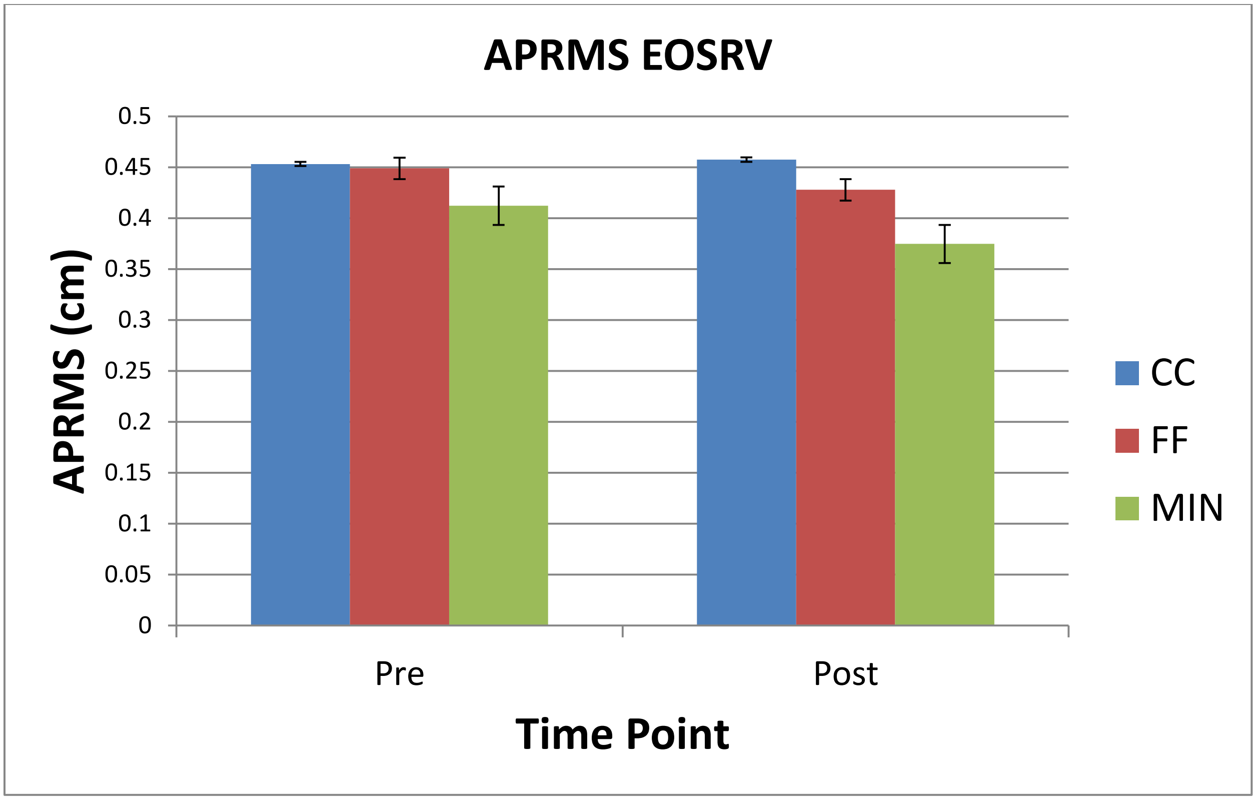
Anterior-Posterior Sway RMS:

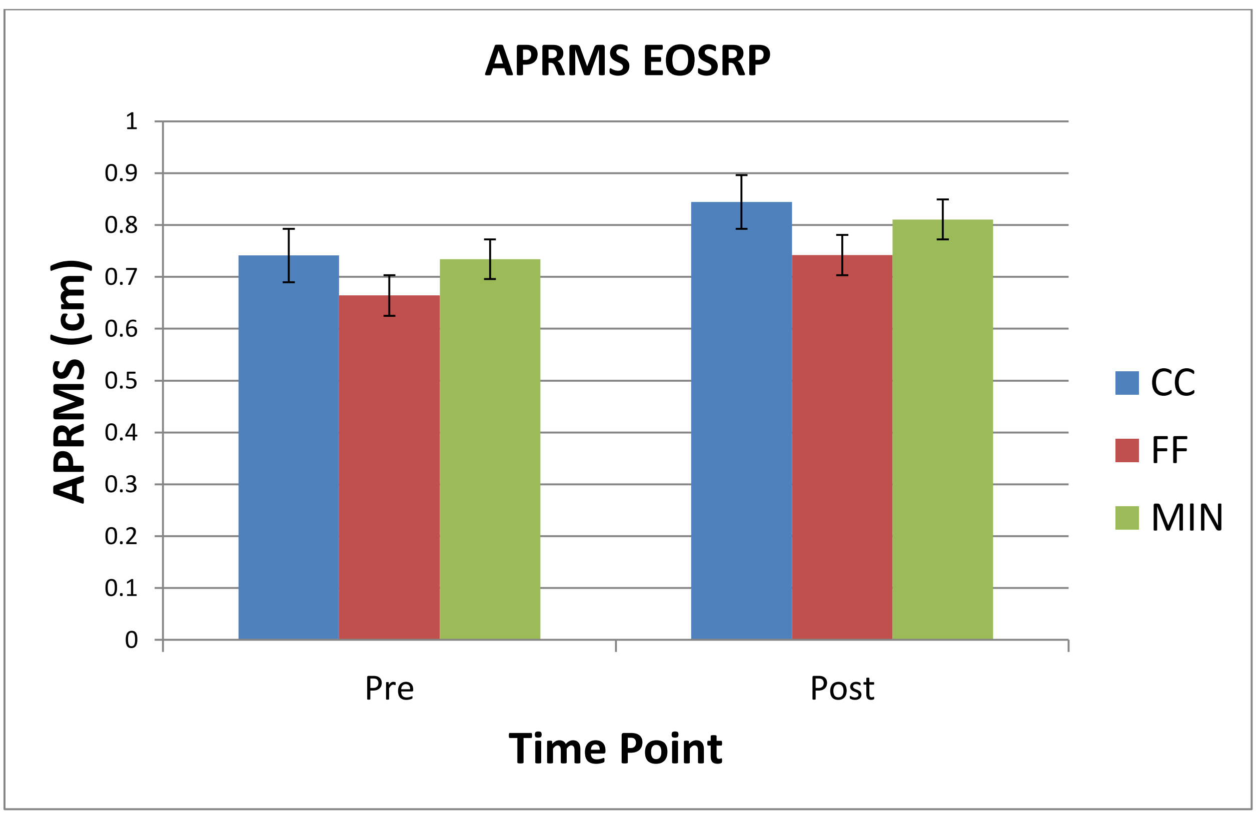
A repeated measures analysis of variance (ANOVA) was used to examine APRMS for each of the four Neurcom® Sensory Organization Test (SOT) conditions. Significant differences were found in condition 2 (EC) for the footwear main effects ($F(2,34) = 7.914$, $p = 0.002$). No significant interaction was seen for the footwear by time variable, thus the APRMS main effects for footwear can be generalized across time points. Pairwise comparisons using a Bonferroni adjustment for multiple comparisons revealed the CC displayed significant differences in APRMS compared to the MIN ($p < 0.0001$). No significant differences for footwear main effects, time main effects, or interaction terms were observed in conditions 1 (EO), 3 (EOSRV), or 4 (EOSRP) for APRMS. These results indicate better balance performance in the minimalist condition.

Figure 3: Averaged Sway RMS measures in the Anterior-Posterior direction for each of the six Neurocom® SOT conditions. * represents a significant difference in footwear conditions, † represents a significant difference across time conditions, # represents a significant interaction and the bars represent the standard error.





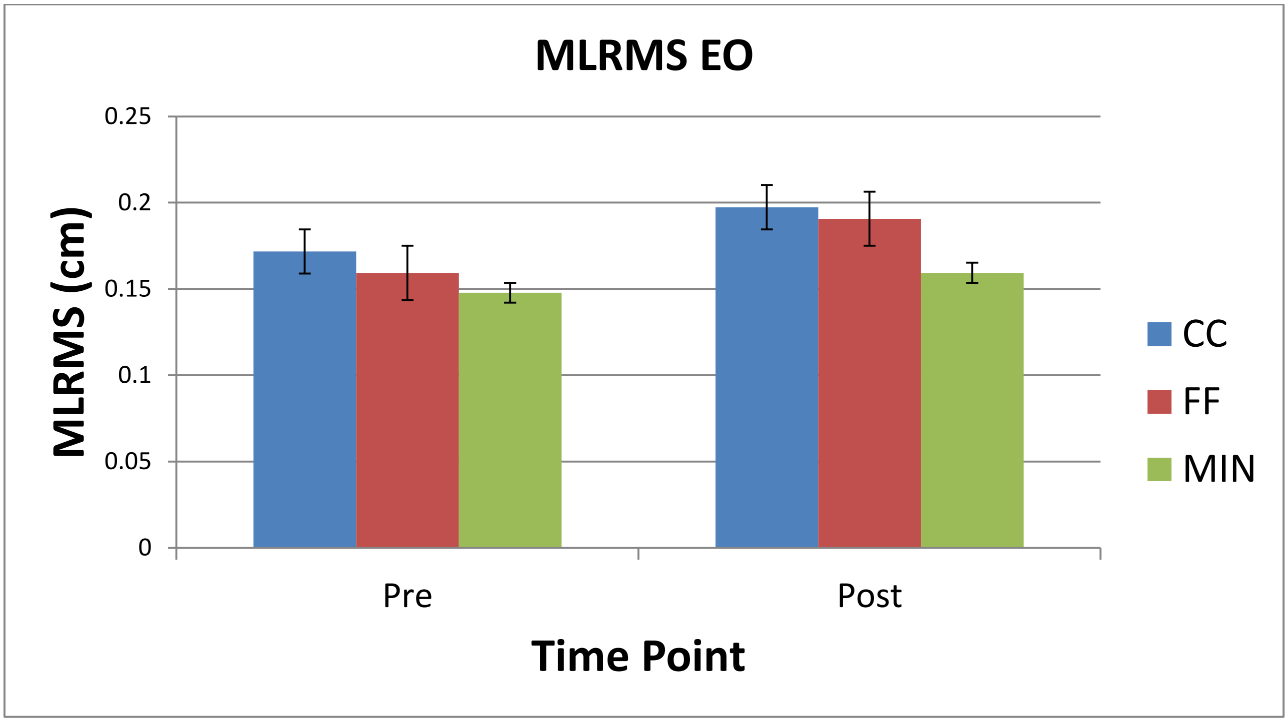

APRMS EOSRV
APRMS (cm)
0.5
0.45
0.4
0.35
0.3
0.25
0.2
0.15
0.1
0.05
0
Pre
Post
Time Point
CC
FF
MIN

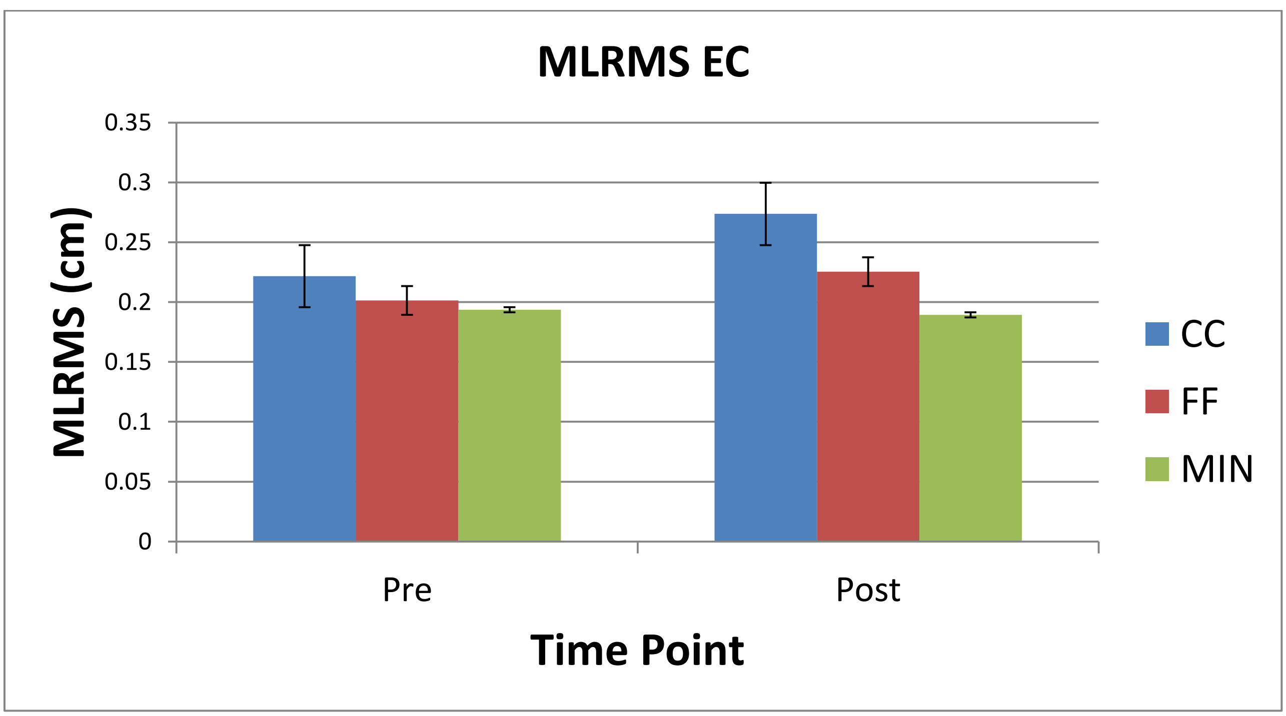

APRMS EOSRP
APRMS (cm)
1
0.9
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
Pre
Post
Time Point
CC
FF
MIN

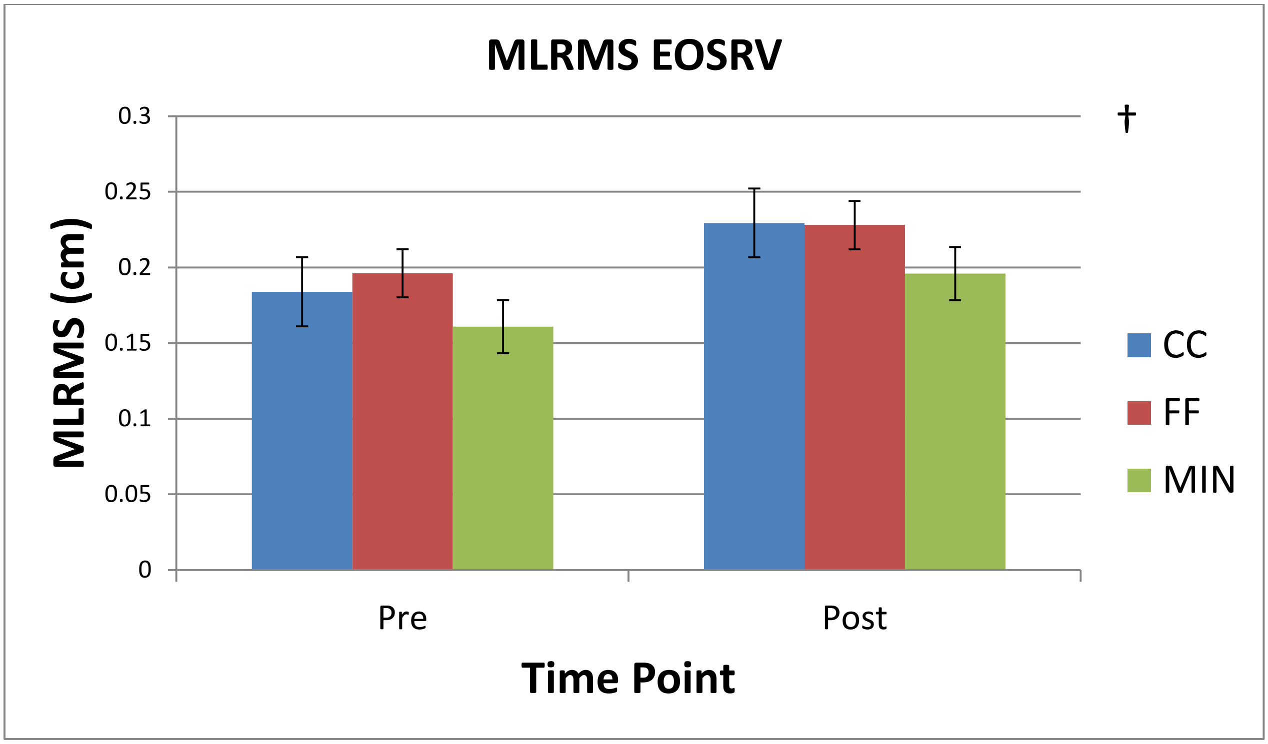
Medial-Lateral Sway RMS:

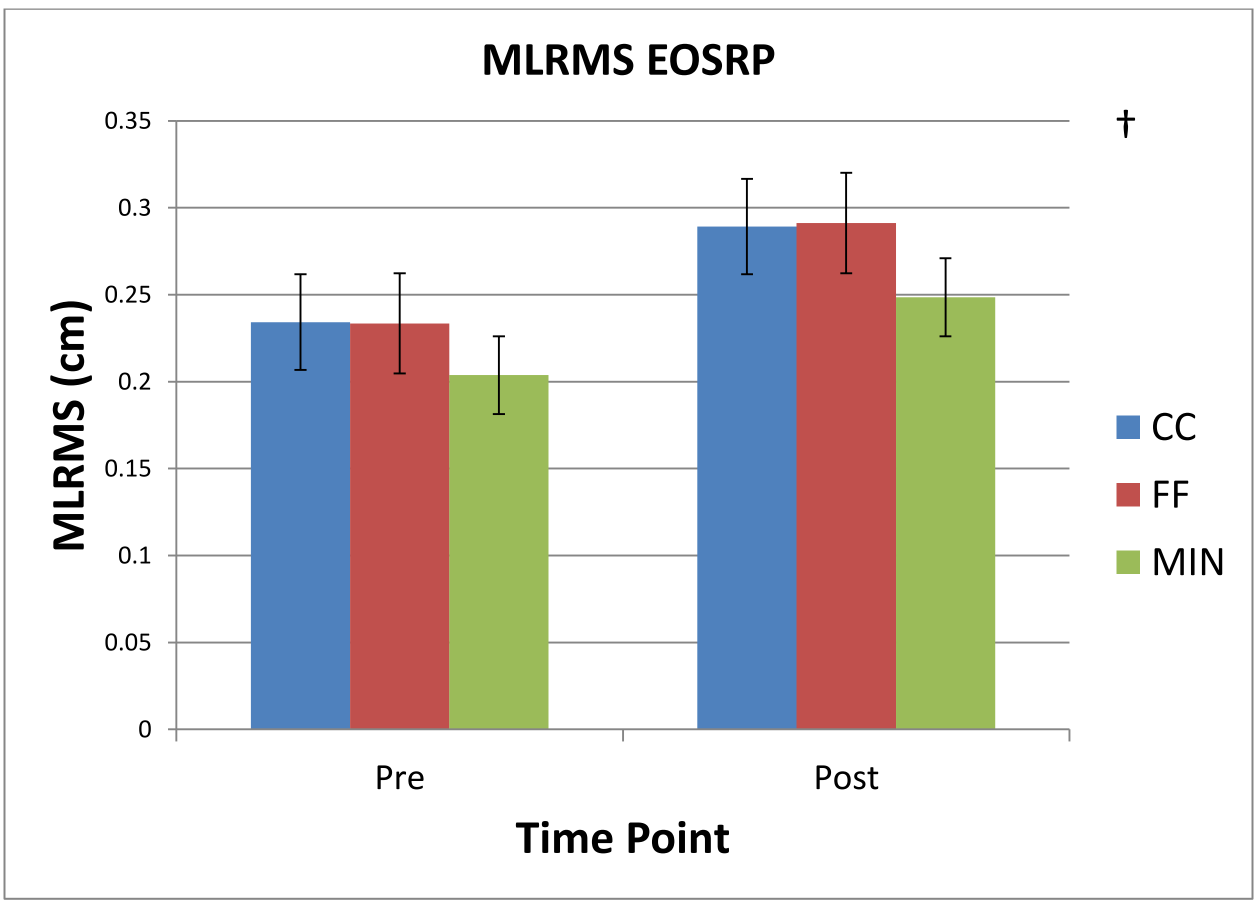
A repeated measures analysis of variance (ANOVA) was used to examine sway RMS in the medial-lateral direction for each of the four Neurocom® SOT conditions. Significant differences were found in condition 3 (EOSRV) for the time main effect ($F(1,17) = 6.532$, $p = 0.02$). No significant interaction effect was found, thus the main effect differences for time can be generalized across footwear types. Significant differences were found in condition 4 (EOSRP) for the time main effect ($F(1,17) = 8.982$, $p = 0.008$. No significant interaction was found in condition 4, thus the main effect differences for time can be generalized across footwear types. No significant differences were found in interaction, footwear main effect or time main effect for conditions 1 (EO) and 2 (EC). These results suggest that lateral sway RMS is increased, and balance potentially declines following the one mile walk.

Figure 4: Averaged Sway RMS measures in the Medial-Lateral direction for each of the six Neurocom® SOT conditions. * represents a significant difference in footwear conditions, † represents a significant difference across time conditions, # represents a significant interaction and the bars represent the standard error.





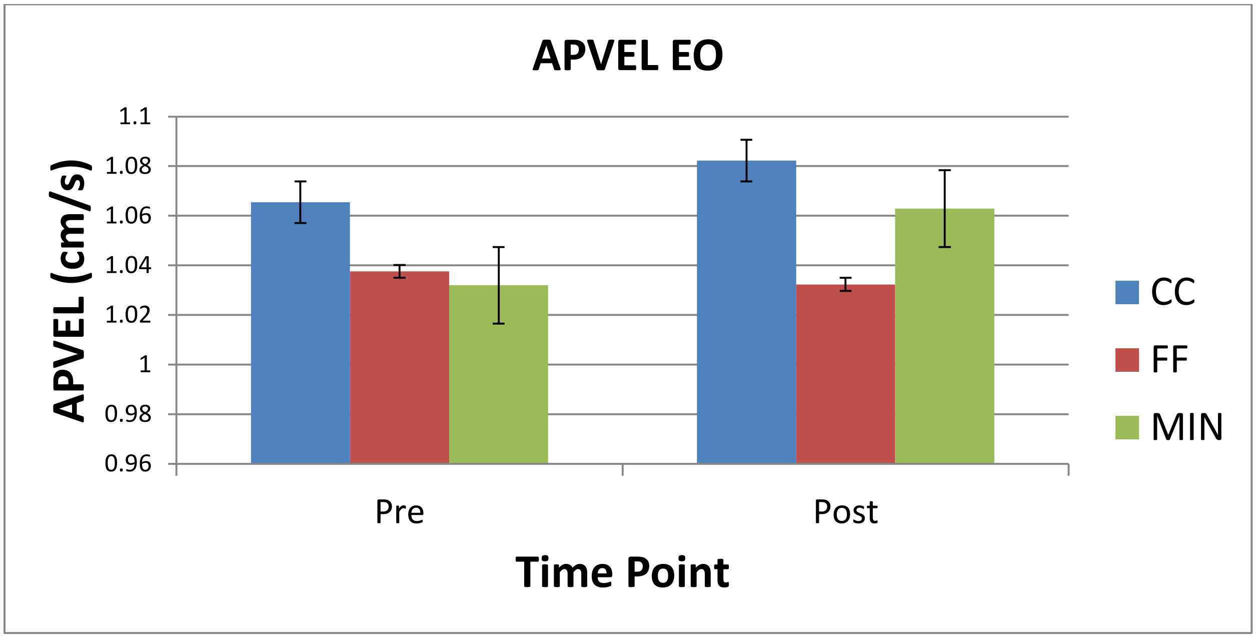

MLRMS EOSRV
0.3
0.25
0.2
0.15
0.1
0.05
0
MLRMS (cm)
†
CC
FF
MIN
Pre
Post
Time Point

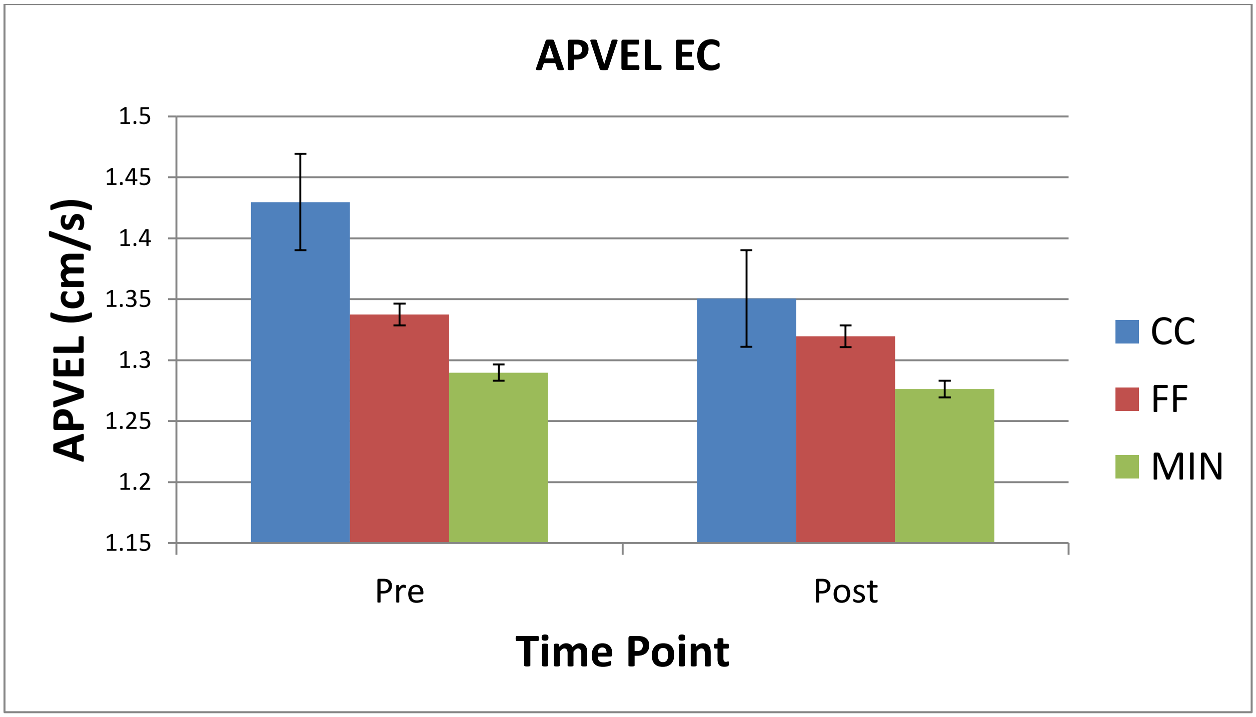

MLRMS EOSRP
0.35
0.3
0.25
0.2
0.15
0.1
0.05
0
MLRMS (cm)
†
CC
FF
MIN
Pre
Post
Time Point

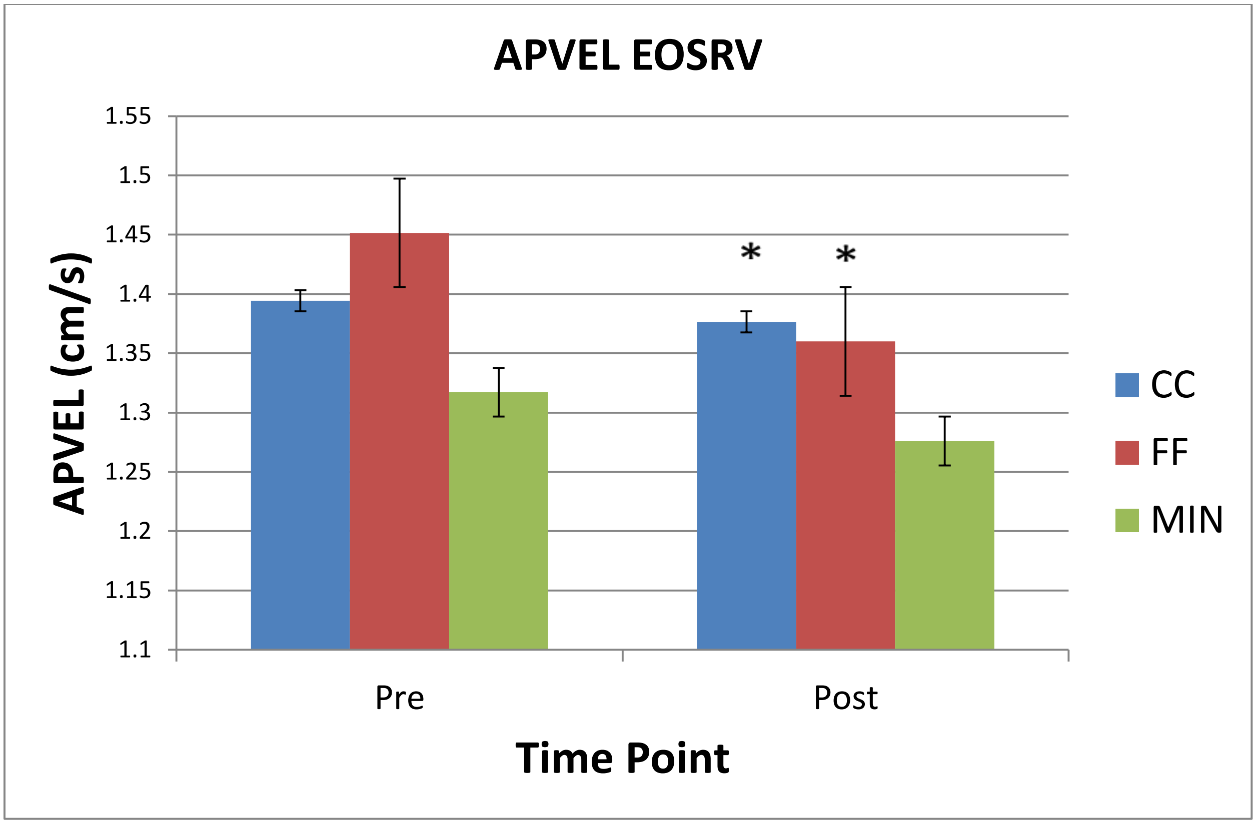
Anterior-Posterior Sway Velocity:

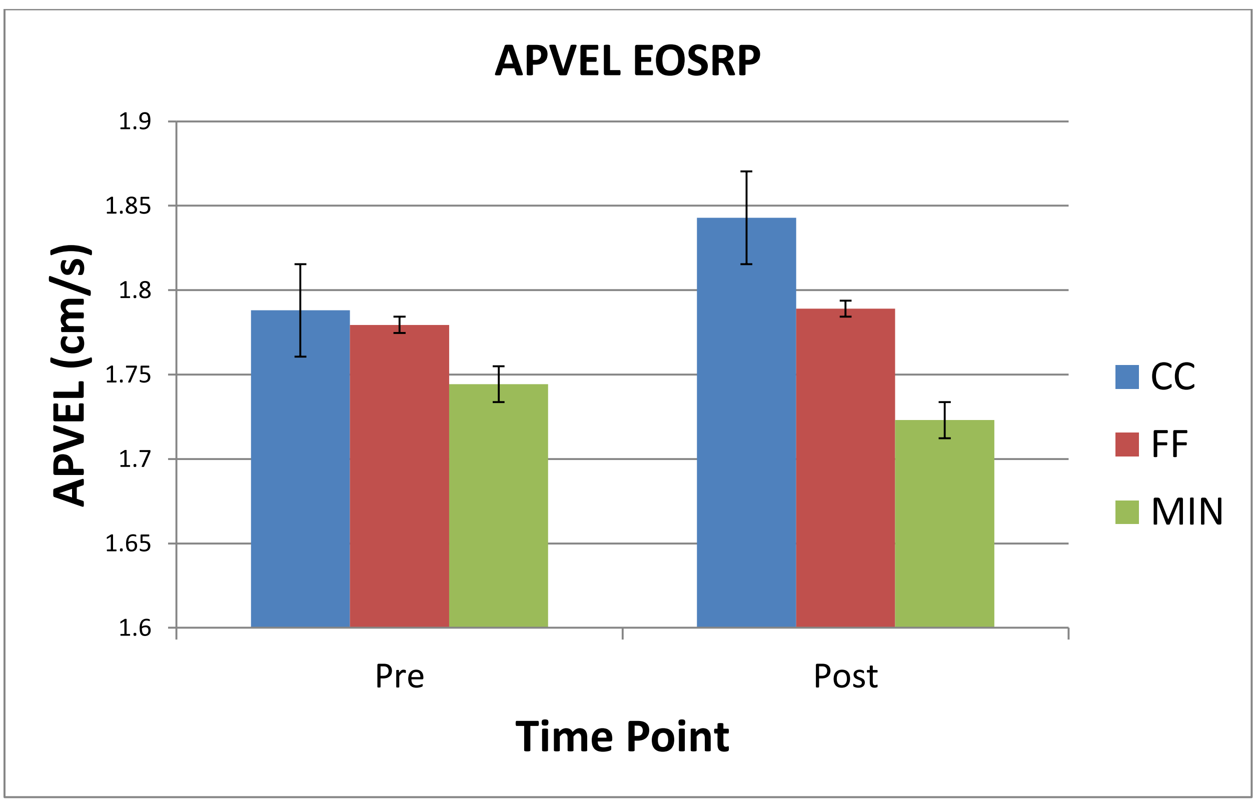
A repeated measures analysis of variance (ANOVA) was used to examine differences in Sway Velocity in the Anterior-Posterior direction for each of the four Neurocom® SOT conditions. Significant differences were found in condition 3 (EOSRV) for the footwear main effect ($F(2,34) = 7.469$, $p = 0.002$). Pairwise comparisons using a Bonferroni correction for multiple comparisons was used. Pairwise comparisons showed significant differences between the MIN and CC ($p = 0.014$) and between the MIN and FF ($p = 0.010$). No significant interaction effect was seen in condition 3, thus the main effect differences for footwear can be generalized across time points. No significant differences for interaction, footwear main effect, or time main effect were found in APVEL for conditions 1 (EO), 2 (EC) and 4(EOSRP).

Figure 5: Averaged Sway Velocity measures in the Anterior-Posterior direction for each of the six Neurocom® SOT conditions. * represents a significant difference in footwear conditions, † represents a significant difference across time conditions, # represents a significant interaction and the bars represent the standard error.





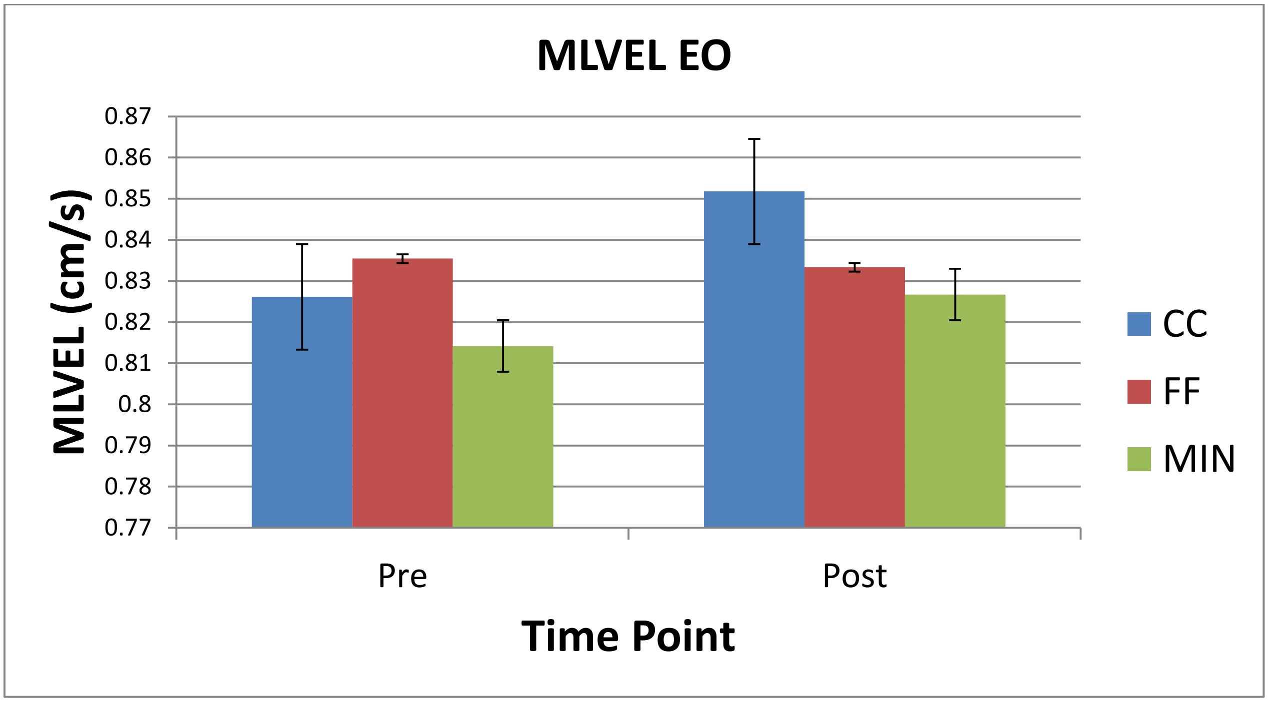

APVEL EOSRV
APVEL (cm/s)
1.55
1.5
1.45
1.4
1.35
1.3
1.25
1.2
1.15
1.1
*
*
Pre
Post
Time Point
CC
FF
MIN

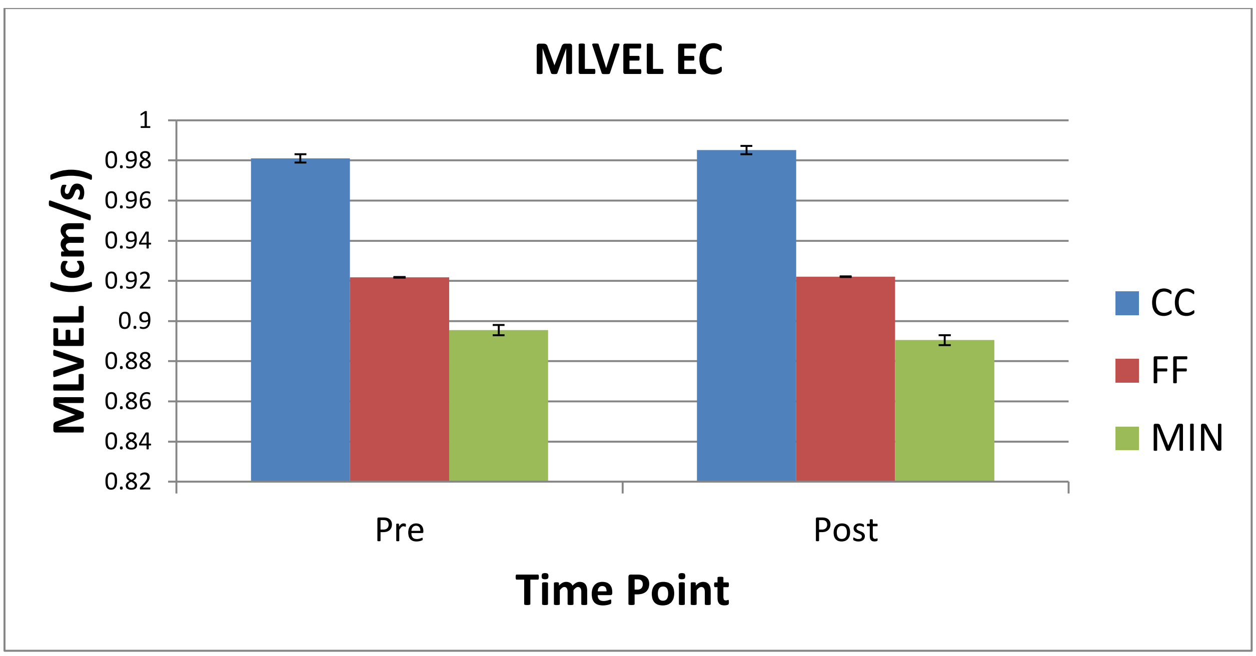

APVEL EOSRP
APVEL (cm/s)
1.9
1.85
1.8
1.75
1.7
1.65
1.6
Pre
Post
Time Point
CC
FF
MIN

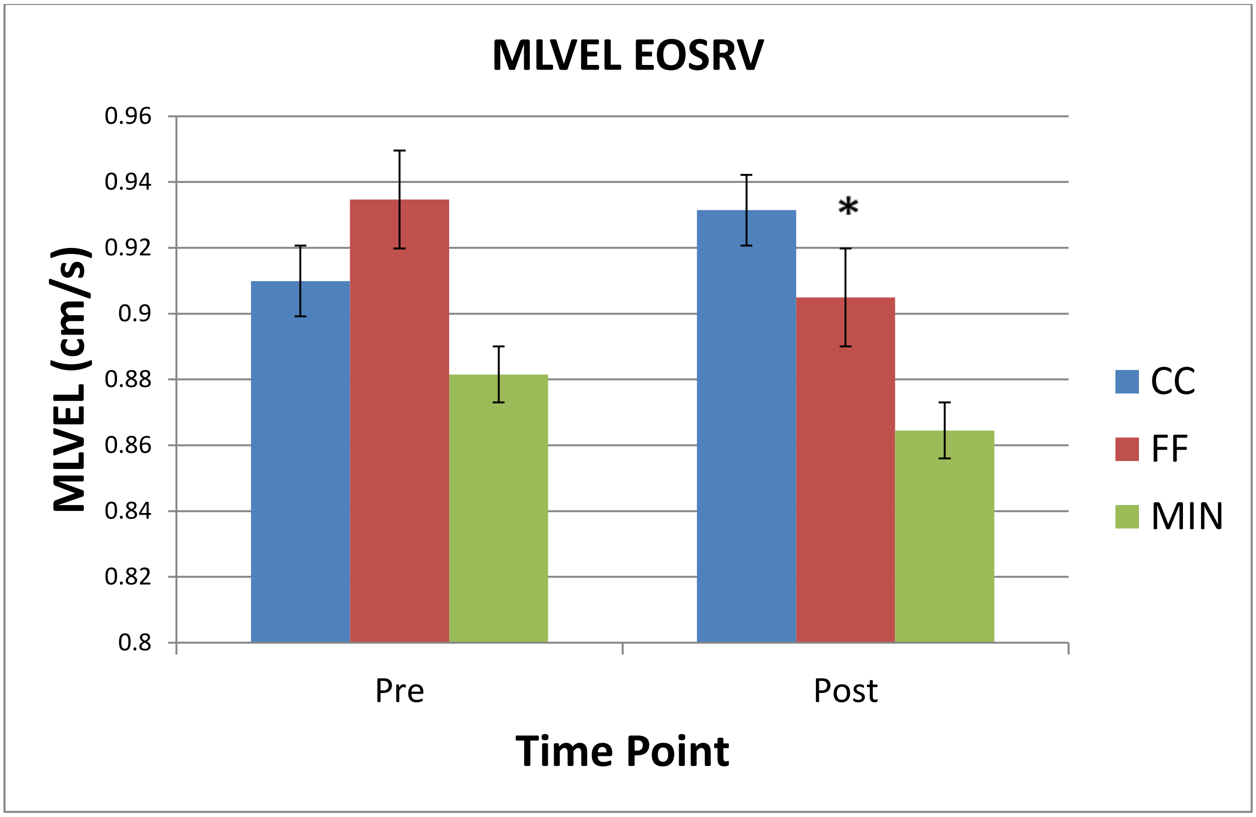
Medial-Lateral Sway Velocity:

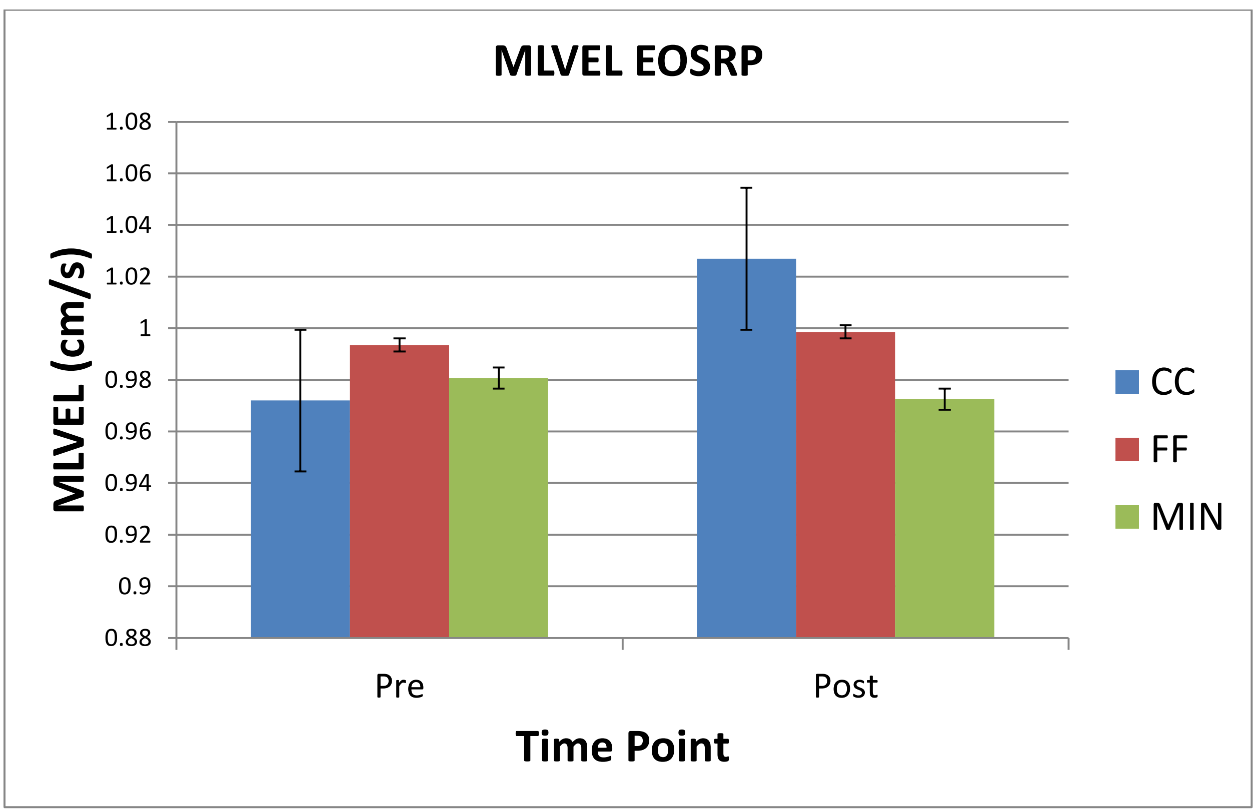
A repeated measures analysis of variance (ANOVA) was used to examine the differences in Sway Velocity in the Medial-Lateral direction for each of the four Neurocom® SOT conditions. Significant differences were found in condition 3 (EOSRV) for the footwear main effect ($F(2,34) = 3.681$, $p = 0.036$). Pairwise comparisons using a Bonferroni correction for multiple comparisons was used. Pairwise comparisons displayed significant differences between MIN and FF ($p < 0.0001$). No significant interaction was seen in condition 3, thus the footwear differences can be generalized across the time points. These results indicate an increase in lateral sway velocity, and potential subsequent decline in postural stability in the flip flop compared to the minimalist.

Figure 6: Averaged Sway Velocity measures in the Medial-Lateral direction for each of the six Neurocom® SOT conditions. * represents a significant difference in footwear conditions, † represents a significant difference across time conditions, # represents a significant interaction and the bars represent the standard error.






MLVEL EOSRV
0.96
0.94
0.92
0.9
0.88
0.86
0.84
0.82
0.8
MLVEL (cm/s)
*
Pre
Post
Time Point
CC
FF
MIN


MLVEL EOSRP
1.08
1.06
1.04
1.02
1
0.98
0.96
0.94
0.92
0.9
0.88
MLVEL (cm/s)
Pre
Post
Time Point
CC
FF
MIN

Reaction Time:

A repeated measures analysis of variance (ANOVA) was used to examine the differences in reaction time latencies (RT) for each of the four conditions BWM, BWL, FWM and FWL) in the Neurocom® motor control test (MCT). No significant differences were found in the BWM condition for footwear main effect ($F(2,34) = 0.359$, $p = 0.701$), time main effect ($F(1,17) = 0.057$, $p = 0.815$) or the interaction term ($F(2,34) = 2.469$, $p = 0.100$). No significant differences were found in the BWL condition for footwear main effect ($F(2,34) = 0.537$, $p = 0.589$), time main effect ($F(1,17) = 0.005$, $p = 0.947$) or the interaction term ($F(2,34) = 0.778$, $p = 0.467$). No significant differences were found in the FWM condition for footwear main effect ($F(2,34) = 1.450$, $p = 0.249$), time main effect ($F(1,17) = 0.064$, $p = 0.804$) or the interaction term ($F(2,34) = 0.526$, $p = 0.596$). Finally in the FWL condition, no significant differences were found for either the footwear main effect ($F(2,34) = 0.691$, $p = 0.508$), time main effect ($F(1,17) = 0.065$, $p = 0.802$) or the interaction term ($F(2,34) = 0.340$, $p = 0.714$).

EMG Analysis:

Mean EMG:

The average EMG was obtained after sampling the raw data at 1,500 Hz, data was filtered using a band-pass (20-250Hz) and full wave rectified. Mean EMG was calculated for each of the three 20 second trials per SOT condition, and then the three trial means were averaged to get the condition mean EMG for the 4 conditions (EO, EC, EOSRV, and EOSRP). A 3 x 2 (footwear x time) repeated measures analysis of variance (ANOVA) was used to examine EMG activity for each SOT condition, in both the medial gastrocnemius and tibialis anterior.

EO:

No significant differences were found for mean EMG of the medial gastrocnemius (MG) for the footwear main effects ($F(2,34) = 0.253$, $p = 0.778$), the time main effects ($F(1,17) = 0.236$, $p = 0.633$), and there was no significance for the interaction term ($F(2,34) = 0.110$, $p = 0.896$). For the tibialis anterior (TA), no significant interaction was observed ($F(2,34) = 0.269$, $p = 0.766$), and no significant differences were seen for the main effects of footwear ($F(2,34) = 0.650$, $p = 0.528$) or time ($F(1,17) = 0.748$, $p = 0.399$).

EC:

In the eyes closed condition (EC), for the medial gastrocnemius (MG) no significant interaction was observed ($F(2,34) = 0.293$, $p = 0.748$). There were also no significant differences observed for the main effects of footwear ($F(2,34) = 0.303$, $p = 0.740$), or time ($F(1,17) = 0.304$, $p = 0.589$). For the tibialis anterior (TA), no significant interaction was observed ($F(2,34) = 1.294$, $p = 0.287$), as well as no significant differences for footwear main effect ($F(2,34) = 1.081$, $p = 0.351$) or time main effect ($F(1,17) = 0.564$, $p = 0.463$).

EOSRV:

No significant interaction was seen for the MG in the EOSRV condition ($F(2,34) = 0.129$, $p = 0.880$) and no significant differences were seen for the footwear main effect ($F(2,34) = 0.576$, $p = 0.567$), or the time main effect ($F(1,17) = 0.159$, $p = 0.695$). Analysis of the TA in this condition yielded similar results, no significant interaction was observed ($F(2,34) = 1.548$, $p = 0.227$) and no significant differences were seen for the main effects of footwear ($F(2,34) = 0.430$, $p = 0.654$) or time ($F(1,17) = 1.228$, $p = 0.283$).

EOSRP:

In the EOSRP condition for the MG, no significant interaction was seen ($F(2,34) = 0.498$, $p = 0.612$), as well as no significant differences for the main effect of footwear ($F(2,34) = 0.071$, $p = 0.932$) or time ($F(1,17) = 0.031$, $p = 0.863$). The TA also showed no significant interaction effect ($F(2,34) = 0.199$, $p = 0.820$), or significant differences for the footwear main effect ($F(2,34) = 0.619$, $p = 0.545$), or time main effect ($F(1,17) = 0.410$, $p = 0.530$).

Muscle Activation:

Prior to all balance testing, each participant completed a series of maximal voluntary isometric contractions (MVIC). Three MVICs were performed for the medial gastrocnemius (MG), tibialis anterior (TA), vastus medialis (Q), and hamstring semitendinosus (H) with a minumum of 30 seconds rest in between trials. Following a band-pass filter (20-250Hz) and full wave rectification, the average EMG was collected for each MVIC, then the three trial means were averaged to obtain the final MVIC for that testing condition . The average EMG was obtained after sampling the raw data at 1,500 Hz, data was filtered using a band-pass (20-250Hz) and full wave rectified. Mean EMG was calculated for each of the three 20 second trials per SOT condition, and then the three trial means were averaged to get the condition mean EMG for the 4 conditions (EO, EC, EOSRV, and EOSRP). These averages were then divided by the condition MVIC obtained earlier, and multiplied by 100 ((Mean EMG/MVIC)*100) in order to obtain a percentage of muscle activation elicited per trial. A 3 x 2 (footwear x time) repeated measures analysis of variance (ANOVA) was used to examine percent activition for each SOT condition, in both the medial gastrocnemius and tibialis anterior.

EO:

The MG in the EO condition showed no significant footwear by time interaction effect ($F(2,34) = 0.786$, $p = 0.464$) and no significant differences for the main effects of footwear ($F(2,34) = 0.159$, $p = 0.853$) or time ($F(1,17) = 0.035$, $p = 0.853$). For the TA no significant interaction effect was observed ($F(2,34) = 0.322$, $p = 0.727$) as well as no significant differences for footwear main effect ($F(2,34) = 0.707$, $p = 0.500$) or time main effect ($F(1,17) = 0.204$, $p = 0.658$).

EC:

In the EC condition for the MG, no significant interaction was seen ($F(2,34) = 2.525$, $p = 0.095$) and no main effect significance for footwear ($F(2,34) = 0.626$, $p = 0.541$) or time ($F(1,17) = 0.128$, $p = 0.725$). The TA muscle activation showed no significant interaction ($F(2,34) = 1.323$, $p = 0.280$) as well as no significant differences for either the footwear main effect ($F(2,34) = 1.013$, $p = 0.374$) or time main effect ($F(1,17) = 0.010$, $p = 0.920$).

EOSRV:

For the EOSRV condition, the MG showed no significant footwear by time interaction ($F(2,34) = 0.048$, $p = 0.953$) and no significant main effect differences for footwear ($F(2,34) = 1.802$, $p = 0.180$) or time ($F(1,17) = 1.927$, $p = 0.183$). The TA showed no significant footwear by time interaction ($F(2,34) = 1.768$, $p = 0.186$) as well as no significant differences for footwear main effect ($F(2,34) = 0.489$, $p = 0.617$) or time main effect ($F(1,17) = 0.926$, $p = 0.349$).

EOSRP:

In the EOSRP condition the MG showed no significant footwear by time interaction ($F(2,34) = 0.486$, $p = 0.620$) and no significant differences for either main effect of footwear ($F(2,34) = 0.119$, $p = 0.888$) or time ($F(1,17) = 0.409$, $p = 0.531$). The TA muscle activation showed no significant footwear by time interaction ($F(2,34) = 0.121$, $p = 0.886$) and no significant differences for footwear main effect ($F(2,34) = 0.104$, $p = 0.902$) or time main effect ($F(1,17) = 0.590$, $p = 0.453$).

Ankle Co-Contraction:

Muscle co-contraction index (CCI) (operationally defined as the simultaneous activation of two muscles) was calculated based on the ratio of the EMG activity of antagonist/agonist muscle pairs of the lower leg between the tibialis anterior, and medial gastroc (TA/MG). Co-contraction was calculated using the following equation (Rudolph, Axe, Buchanan, Scholz, and Mackler, 2001).

(EMGS/EMGL) x (LowerEMG + HigherEMG)

Where EMGS was the level of activity in the less active muscle, EMGL was the level of activity in the more active muscle. This ratio was multiplied by the sum of the activity found in the two muscles. This method has been used because it provides an estimate of the relative activation of the pair of muscles as well as the magnitude of the co-contraction. Co-contraction was calculated for all three trials of each SOT condition, and the three trials were averaged to obtain one CCI per SOT condition.

EO:

Co-contraction at the ankle in the EO condition showed no significant footwear by time interaction ($F(2,34) = 0.510$, $p = 0.605$), and no significant differences for the main effect of footwear ($F(2,34) = 0.697$, $p = 0.505$) or time ($F(1,17) = 0.051$, $p = 0.824$)

EC:

In the EC condition ankle co-contraction showed no significant footwear by time interaction ($F(2,34) = 1.451$, $p = 0.248$), as well as no significant differences for footwear main effect ($F(2,34) = 1.054$, $p = 0.360$) or time main effect ($F(1,17) = 1.315$, $p = 0.267$).

EOSRV:

Ankle co-contraction in the EOSRV condition showed no significant footwear by time interaction ($F(2,34) = 2.590$, $p = 0.090$) or significant differences in main effects for footwear ($F(2,34) = 0.760$, $p = 0.476$) or time ($F(1,17) = 0.012$, $p = 0.914$).

EOSRP:

Finally, in the EOSRP condition no significant footwear by time interaction was seen for ankle co-contraction ($F(2,34) = 1.210$, $p = 0.311$), and no significant differences were seen for the main effects of footwear ($F(2,34) = 0.942$, $p = 0.400$) or time ($F(1,17) = 0.001$, $p = 0.998$).

CHAPTER 5

DISCUSSION

The purpose of this study was to examine the effects of three types of popular casual footwear, including the clog style Croc® (CC), thong-style flip-flop (FF), and the Vibram® Five-Finger minimalist (MIN) on the human postural control system following a one mile walk. Balance measures were recorded using four of the conditions of the Neurocom® Equitest (EO, EC, EOSRV, and EOSRP). The average sway velocity and route-mean-square (RMS) of the center of pressure (CoP) were used to quantify the postural sway in the anterior-posterior (APVEL & APRMS) and the medio-lateral (MLVEL & MLRMS) directions. This study hypothesized that differences in balance performance would be observed across time points, as well as between footwear conditions. Significant differences were found for both main effects of time and footwear in select sway parameters during the EC, EOSRV, and EOSRP conditions of the SOT.

Balance Performance as a Function of Time

There is previous evidence to suggest that balance performance may decline over extended time as a result of being on your feet (Chander, Garner, & Wade, 2014; Wade & Davis, 2008), with marked declines reported after the second hour (Cham & Redfern, 2001). The results seen in this study agree with those previously in the literature. Significant differences across time points were observed in conditions 3, and 4 of the Neurocom® Sensory Organization Test (SOT). Similar results have been seen in occupational footwear (Cham & Redfern, 2001;

Chander et al., 2014; Wade, Garner, Redfern, & Andres, 2013). Interestingly, results seen currently are observed after a much shorter time duration, suggesting that a potential decline in somatosensory feedback accuracy could be observed after only a one mile walk. The differences in findings between time points could also be attributable to the methodology employed. The aforementioned studies, while using a much longer time span, allowed participants to stand/walk around at a self-selected pace, while also taking 1-2 minute breaks throughout the testing period (Chander et al., 2014; Wade & Davis, 2008), or were standing at a control panel performing cognitive tasks and stopped to walk for 2 minute intervals (Cham & Redfern, 2001). In contrast to this study in which after the preferred walking speed was determined, the one mile was walked in its entirety, with post testing immediately. For the current study, time point comparisons showed that in the EOSRV, and EOSRP conditions, MLRMS in the post testing was significantly higher than that of the pre testing. This is particularly of interest because while the current results are observed in younger individuals, the footwear examined is not independent to this age group and may be worn by all ages. It has been previously noted in the literature that increased amounts of postural sway, may ultimately lead to falls in older adults (Fernie, Gryfe, Holliday, & Llewellyn, 1982) and more specifically, MLRMS is the leading predictor of increased fall risk in elderly populations (Maki, Holliday, & Topper, 1994; Mitchell, Collins, De Luca, Burrows, & Lipsitz, 1995). Laughton et al. (2003) showed that elderly adults that had experienced self-reported falls, showed significant increases in anterior-posterior sway compared to younger individuals. Further, Laughton et al. made comparisons between sway parameters and elderly participants' scores on the performance oriented mobility assessment (POMA), a clinical balance assessment. These results indicated that elderly subjects who performed poorly on the POMA also displayed significant increases in medial-lateral postural sway, while those who scored higher on the POMA did not show these increases (Laughton et al., 2003). While the results here are shown in a young population it is only speculatory that the results would manifest in elderly populations in a similar way. Due to age related effects, it is possible that balance in the elderly population would show exacerbated effects. With advancing age, the afferent response of muscle spindles to varying levels of stretch appears to decline. Showing lower discharge rates and potentially a decline in spindle sensitivity. Morphological changes such as increased capsular thickness and a decreased number of intrafusal fibers may account for the dampening of static and dynamic muscle spindle sensitivity that is seen with aging (Shaffer & Harrison, 2007). However, because of the increasing importance of fall prevention, future studies should further investigate these findings in an elderly population.

Balance Performance as a Function of Footwear

Previous literature has suggested that footwear properties can affect balance performance by adversely affecting the body's sensory systems (Brecht, Chang, Price, & Lehmann, 1995; Chander et al., 2014; Ko, & Lee, 2013; Menant et al., 2008; Ottaviani, Ashton-Miller, Kothari, & Wojtys, 1995; Robbins, Waked, Gouw, & McClaran, 1994; Rose et al., 2011). It is commonly reported in the literature that properties such as mass (Pline, Madigan, & Nussbaum, 2006; Chander et al., 2014; Garner, Wade, Garten, Chander, & Acevedo, 2013), sole thickness (Robbins et al., 1994; Menant et al., 2008), heel height (Menant et al., 2008; Brecht et al., 1995) and heel-hold (Lindemann et al., 2003) all may play some role in postural control. A study done by Robbins (1994) examined the effects of midsole hardness and midsole thickness on balance performance in men. The results of Robbins and colleagues (1994) suggested that a thinner, harder midsole was the best for maintaining balance while walking on a balance-beam. The results from the current study observed significant increases in postural sway parameters between the Croc® and minimalist conditions, representing the thickest/softest, and thinnest/hardest soles, respectively. However, previous reports have suggested that no changes in stability are observed until heel heights reach that greater than 4 cm (Menant et al., 2008). This suggests that if the differences between the Croc® and minimalist conditions are due to sole characteristics, that the hardness may attribute more to these differences than thickness. However, statistical differences in sole hardness were also seen between the Croc® and flip-flop, which showed no significant differences in balance. This suggests that combinations of footwear characteristics, and not one footwear characteristic alone may be contributing to balance decrements. It was also hypothesized that the properties of the footwear collar and how it secured to the foot would affect balance performance. The Croc®, while not directly securing to the ankle, was worn with the strap behind the heel providing some support in keeping it attached to the foot. The flip-flop's only attachment to the foot being the thong actively held between the toes, and the minimalist completely secured to the ankle. Previous literature has suggested that footwear with a firm heel-hold (defined previously as slippers or footwear with cut-away heels) (Gabell, Simons, & Nayak, 1985; Lindemann et al., 2003) may improve balance performance by providing additional sensory feedback, as well as requiring less muscular activity in order to actively hold the footwear on the foot compared to that of slippers or flip-flops (Lindemann et al., 2003). While that notion is supported by the decreased amount of postural sway in the minimalist condition, and by an increase in postural sway parameters when participants donned the Croc®. The flip-flop showed a decrease in postural sway in the post testing. Previous literature has shown decreased dorsiflexion in flip flops when compared to sneakers. This was concluded to be due to increased activity of the Flexor Hallucis Longus (FHL) and flexor digitorum longus (FDL) to flex the phalanges to grip the flip-flop during the swing phase in an attempt to keep the flip-flop on the foot. This increased activity of the toe flexors would subsequently cause an increased plantar flexion moment at the ankle reulting in decreased dorsiflexion, and a concurrent increase in the dorsiflexor muscles to counter act the increased plantar flexion moment (Shroyer, 2009). This increase in EMG activity over time may have potential effects of fatigue in the lower limb musculature and proprioceptors, and a subsequent decline in afferent information. Further, previous literature has suggested that when proprioceptive feedback at the ankle becomes less reliable due to age related effects, there is an adoption of a new co-contraction strategy for maintaining postural stability (Benjuya, Melzer, & Kaplanski, 2004). However, based on the findings of the current study, we did not see any significant increases in EMG activity of the lower limb, or any significant co-contraction. Thus, we do not have any evidence to suggest that there is fatigue in the lower limb. Finally, footwear mass is commonly associated with changes in postural sway when worn for extended durations. It has been suggested that an increased footwear mass is associated with an increase in energy expenditure, and subsequent increase in fatigue when worn for extended durations (Chander et al., 2014; Garner et al., 2012; Jones, Toner, Daniels, & Knapik, 1984). Previous reports indicate that there is an increase in energy expenditure of 0.7 – 1.0% for every 100g of added footwear mass (Jones et al., 1984). The previous literature suggesting that mass played a key role in balance decrements showed a 0.5 kg difference between firefighter boots (2.4kg and 2.9 kg) (Garner et al., 2012), and between occupational footwear (0.4kg, 0.5kg, and 0.9kg) (Chander et al., 2014). The current footwear not only have less mass (CC = 0.4kg, FF = 0.2kg, and Min = 0.3kg), but a smaller difference between them. To further suggest that mass is not a key attribute to differences currently, all of the current footwear showed statistically significant differences in mass. However, stability differences were observed between the CC and Min conditions, which were the most similar in mass (396g vs 340g, respectively), and interestingly, no differences were observed between the CC and FF condititions, which had the biggest difference in mass (396g vs 226g, respectively). Based on the different characteristics shown in the current footwear, we cannot elucidate one specific footwear characteristic responsible for the differences in balance performance following a one mile walk. It appears that combinations of these characteristics are likely associated with postural instability. One possible combination is the sole/midsole hardness (CC = Shore A51.2, FF = Shore A71.4, and Min = Shore A99), and a secure heel hold. Previous literature has shown increases in postural instability with increased mass and over extended durations (Chander et al., 2014; Garner et al., 2013). Garner et al. examined the effects of two different types of fire fighter boots on postural control. The primary difference between the two boots being the mass, fit, and support based on different material builds. Their findings showed significant differences between boot types, which they attributed to the increased mass of the rubber boot compared with the leather, as well as before and after a simulated fire stair climb (SFSC) was conducted while wearing an addition 75 lbs in order to simulate the weight of protective clothing and high-rise pack. The SFSC is a standard testing protocol incorporated in the Candidate Physical Ability Test (CPAT) . These results were hypothesized to be due to the increased mass having an effect on fatigue in the long-term, and sensory feedback in the short-term. However, based on the smaller masses seen in the current footwear, and the lack of differences in balance performance across the greatest to least mass, it doesn’t appear that mass is a major contributor to the increases in postural sway observed. This could be due to the aforementioned reports of energy expenditure increasing approximately 0.7-1.0% per 100g, it suggests that the difference in mass between the current footwear of approximately 114g to 170g may cause an approximate 1-2% increase in energy expenditure over a one mile walk, which may not be adequate enough to cause postural deficits. As a function of sole hardness, significant differences in balance between the Croc® and minimalist support the idea that a softer sole/midsole has negative effects on postural sway. However, with the flip-flop and Croc® also being statistically different in hardness and not showing balance differences, it suggests hardness alone may not be a contributing factor. The flip-flop compared to the minimalist showed marked increases in postural sway. This could be due to the effect of heel-hold. The Croc® was worn by participants with the heel strap in place, thus while not providing a direct heel-hold, it did provide some indirect hold. The flip-flop however, lacked heel-hold completely. A possible explanation for why we did not observe differences in postural sway between the flip-flop and Croc® is the distinct characteristics of these footwear. The lack of heel-hold in the flip-flop, and softer sole of the Croc® may have caused similar increases in postural sway. Interestingly, with regards to the increased sway seen in the Croc®, this increase was seen in the absence of accompanied EMG activity increase. It has been suggested that quiet standing postural control utilizes both an open and closed loop system, in the short term and long term, respectively. The open loop control acts in the absence of feedback, primarily through reflex activity of muscle spindle afferents (MSA) (Mitchell, Collins, De Luca, Burrows, & Lipsitz, 1995). The accuracy of the MSA may decline due to fatigue (Pederson, jubisavljevic, Bergenheim, & Johansson, 1998; Sharpe & Miles, 1992), or because of changes seen with advancing age (Shaffer & Harrison, 2007). However, based on our findings we can only speculate that the low workload of the walk was enough to alter the accuracy of MSA, possibly due to the increased mass of the Croc®, while not being a high enough workload to cause a shift to a closed loop control system. This could explain the increase in sway due to declines in MSA accuracy and firing rates, without the increase of volitional muscle activity seen in EMG. The minimalist footwear displayed significantly greater postural stability compared to other footwear conditions as well as over time. These observations may be based on footwear properties of the minimalist, and its design to mimic barefoot conditions. The primary characteristics of the minimalist that appear to provide improvements in stability are the thinner, hard sole, and secure heel-hold. It is possible that the decreased mass attributed as well, but based on the increased instability in the flip-flop which had the lowest mass of the three footwear conditions. It may indicate that the footwear mass is not the main component. However, future studies should examine footwear comparisons with similar heel-hold properties of varying mass to confirm this hypothesis.

Reaction Time Latencies

Previous literature has showed delayed postural response latencies in response to the motor control test (MCT) while wearing flip-flops and occupational footwear and increases in latency measures in barefoot conditions (Chander et al., 2013; Hosoda et al., 1998). The differences observed in those studies was attributed to the material characteristics of the footwear. Hosoda et al. (1998) showed significant differences between different types of sandals in latencies. The sandals included slippers with and without thongs, leather soled sandals, and Japanese socks. Three of the four types shared the thong characteristic. Their results indicated that the sandals with thongs showed significantly faster latencies, which was attributed to mechanoreceptor stimulation due to the separation of the toes (Hosoda et al., 1998), however cultural differences may have had an effect as well. The results from the current study showed no statistically significant differences in latency responses between footwear or time conditions. The contrasting findings could be due to the design features of the types of casual footwear used, or the physiological workload of the self selected pace one mile walk may not have been fatiguing and sufficient to cause changes in the postural response latencies during external balance perturbations in college aged healthy male adults. Future research on these types of casual footwear should utilize increased physiological workloads in order to better understand the efficiency of these footwear and their effects on the postural control system.

Limitations

There are some obvious limitations within this study that should be noted. First, the lack of a barefoot condition. While this may lead to some problems in inference of results due to lack of a control group, the differences observed from the time the participants first donned the footwear to the end of the testing condition cannot be overlooked in their significance. Second, it could be argued that based on some participants former use of the footwear used, which we did not screen for, the footwear could have been novel for some participants but not for others. However, based on previous literature, evidence suggests that 5 weeks' habituation to new senior shoes does not significantly affect standing balance or gait in older women (Lindemann et al., 2003; Menant, Steele, Menz, Munro, & Lord, 2008).

Conclusions

The purpose of this study was to examine the effects of different types of casual footwear on human balance performance following a one mile walk. The results from this study found significantly increased postural instability between the Croc®, flip-flop, and minimalist conditions. The minimalist footwear exhibited superior balance performance compared to the other footwear conditions across time points. Results from this study suggest that the individual footwear characteristics may not be as associated with increased instability, but certain combinations of these characteristics may be more likely attributable to the balance decrements. Based on these findings, future studies should examine which combinations of footwear characteristics such as sole/midsole thickness and hardness, mass, and heel-hold may be contributing the most to changes in postural stability in order to aid in footwear design that utilizes the optimal combinations of these parameters for the safety of the wearer.

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VITA

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Educational Training:

2012-2015	Master of Science, Exercise Science, May 2015
	The University of Mississippi, University 38677
	Thesis Title: The Influences of Casual Footwear on Human Balance.
2010-2012	Bachelor of Science, Exercise Science, Summer 2012
	The University of Mississippi, University 38677

Employment History:

2012-Present	Graduate Teaching Assistant, Department of Health, Exercise Science, and Recreation Management, University of Mississippi, Oxford MS

Research Experience:

2013-Present	University of Mississippi Applied Biomechanics Laboratory
	Laboratory Associate (2013-Present)

Curriculum Experience:

2014-Present ES 338: Motor Learning and Control (Fall 2014, Spring 2015)

University of Mississippi

Course Description: The study and analysis of motor learning and control with emphasis on sensory, perceptual, and muscular components of human movement.

2014-Present ES 349: Physiology of Exercise Laboratory (Fall 2014)

University of Mississippi

Course Description: Study of body systems affected by exercise; functions of these systems during exercise; effects of age, body type, and nutrition on capacity for exercise; techniques of assessing physical work capacity, and a critical analysis of research literature

2014-Present ES 447: Biomechanics of Human Movement Laboratory (Summer 2014)

University of Mississippi

Course Description: Quantification of the forces acting on the human body during selected activities.

2014-Present ES 457: Exercise Testing & Prescription Laboratory (Spring 2014, Summer 2014)

University of Mississippi

Course Description: Scientific foundations of conditioning, a life-span approach with specific attention to exercise prescription

2013-Present HP 203: First Aid and CPR (Fall 2013, Spring 2014, Summer 2014)

University of Mississippi

Course Description: Safety instruction and practices in the methods as prescribed in the American Red Cross standard and advanced courses.

2013-Present EL 156: Jogging (Fall 2013)

University of Mississippi

2014-Present EL 151: Weight Lifting (Spring 2014)

University of Mississippi

2013-2014 ES 446: Biomechanics of Human Movement (Guest Lecture) (Summer 2014)

University of Mississippi

Course Description: Quantification of the forces action on the human body during selected activities

Abstracts/Oral Presentations:

1. Wilson SJ, Chander H, Morris CE, Wade C & Garner JC. Alternative Footwear's Influence on Static Balance Following a One Mile Walk. SEACSM Annual Meeting, Jacksonville, FL, February 12-14, 2015

2. Chander H, Morris CE, Wilson SJ, Knight AC, Holmes ME, Wade C & Garner JC. Impact of Alternative Footwear on Balance Perturbations. SEACSM Annual Meeting, Jacksonville, FL, February 12-14, 2015.

3. DeBusk H, Chander H, Morris CE, Wilson SJ, Knight AC, Holmes ME, Hill CM, Wade C & Garner JC. The Effect of Commonly Used Alternative Footwear on Balance. SEACSM Annual Meeting, Jacksonville, FL, February 12-14, 2015.

4. Hill CM, Chander H, Morris CE, Wilson SJ, Knight AC, Holmes ME, DeBusk H, Wade C & Garner JC. Impact of Low Intensity Workload on Muscle Exertion in Alternative Footwear. SEACSM Annual Meeting, Jacksonville, FL, February 12-14, 2015.

5. Wilson SJ, Chander H, Morris CE, Garner JC, & Wade C. Alternative Footwear’s Influence on Static Balance Following a One Mile Walk. ACSM Annual Meeting, San Diego, CA, May 26-30, 2015.

6. DeBusk H, Chander H, Morris CE, Wilson SJ, Knight AC, Holmes ME, Hill CM, Wade C & Garner JC. The Effect of Commonly Used Alternative Footwear on Balance. ACSM Annual Meeting, San Diego, CA, May 26-30, 2015.

7. Garner JC, Chander H, Morris CE, Wilson SJ, & Wade C. Impact of Alternative Footwear on Balance Perturbations. ACSM Annual Meeting, San Diego, CA, May 26-30, 2015.

8. Hill CM, Chander H, Morris CE, Wilson SJ, Knight AC, Holmes ME, DeBusk H, Wade C & Garner JC. Impact of Low Intensity Workload on Muscle Exertion in Alternative Footwear. ACSM Annual Meeting, San Diego, CA, May 26-30, 2015.

9. Wilson SJ, Chander H, Morris CE, Garner JC, Waddell DE & Wade C. Alternative Footwear's Influence on Muscle Activation Patterns of the Lower Leg Following a One Mile Walk. Abstracted: *Proceedings of the American Society of Biomechanics Annual Meeting*, Columbus, OH, August 5-8, 2015.

PROFESSIONAL AFFILIATIONS

2013-Present American Red Cross
2015-Present American College of Sports Medicine (ACSM)
2015-Present American College of Sports Medicine (Southeast Regional Chapter) (SEACSM)

LABORATORY COMPETENCIES

1. Exercise Prescription
2. Muscular Strength Testing
3. Extensive Weight Training Experience
4. Electromyography
5. VO2max Testing
6. Vicon Motion Capture Testing
7. Neurocom Equitest Balance Testing
8. Data entry/statistical analysis (SPSS)
9. First Aid/CPR/AED Certified (American Red Cross)
10. First Aid/CPR/AED Instructor Certified (American Red Cross)

Services:

2014-2015	University of Mississippi RebelWell Wellness on Wheels Health Fair event coordinator
2014-Present	Graduate Student Counsel Senator University of Mississippi

MENTORSHIP

2012-Present	John C Garner, III, PhD, CSCS