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Investigating the effects of alternative footwear on balance

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INVESTIGATING THE EFFECTS OF ALTERNATIVE FOOTWEAR ON BALANCE

By David May

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ABSTRACT

Falls are one of the leading causes of injuries and unintentional deaths in the United States, with 27,800 fatalities attributed to falls in 2012 (National Safety Council, 2014). With two thirds of our body mass located two thirds of our body height above the ground, humans require constant work from balance control systems to prevent these falls (Winter, 1995). Because shoes can alter somatosensory input from the mechanoreceptors on the bottom of the foot and affect these balance control systems (Menant et al, 2008), they must be taken into account when looking into the causes of falls. Traditional footwear designs have been studied extensively to examine which shoe characteristics are best for stability. However, with the recent advent of alternative styles of footwear, more research is needed to determine how these new styles affect balance control. The purpose of this study was to determine how three of these types of footwear (Crocs®, flip flops, and Vibram® Five Fingers) affect postural control in 18 healthy males between the ages of 18-44 after walking one mile at a self selected pace. The specific aims of the study were to (1) investigate the effects of a one mile, preferred pace walk on standing balance measures and to (2) examine the effects of three alternative styles of footwear on standing balance measures. These standing balance measures included AP/ML (Anterior-Posterior/Medial-Lateral) sway velocity and AP/ML sway RMS (root-mean-square). Sway velocity and RMS were measured under four conditions of the Neurocom Equitest Sensory Organization Test (EO, EC, EOSRV, EOSRP) before and after the one mile walk. Higher values for sway velocity and RMS were used to indicate decreased balance and postural stability. A predetermined alpha level of 0.05 was used, and results were analyzed using a 2x3 repeated measures ANOVA [2 measurement times (pre/post walking) x 3 footwear types (CC, FF, MIN)] for each of the four sensory organization test (SOT conditions).

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CHAPTER I

INTRODUCTION

Proper functioning of balance control systems is obviously crucial for carrying out the daily activities of life. Because of a relatively high center of mass and a bipedal orientation, humans are inherently unstable without these balance control systems (Winter, 1995). The body's center of mass must be constantly maintained over a continuously changing base of support in order to retain postural control (Winter, 1995). Three major sensory systems work together to provide the central nervous system with the information needed to counteract balance perturbations. The visual, vestibular, and somatosensory systems all help detect changes in the environment that could lead to a fall. The visual system uses the eyes to provide feedback about changing environmental conditions and the body's position in the environment. The somatosensory system utilizes a variety of sensory organs such as the golgi tendon organ, muscle spindles, joint receptors, proprioceptors, and sensory receptors on the foot sole (Levangie, P.K. and Norkin, C.C. 2006) to provide information about external stimuli and the orientation of the body's joints. The vestibular system relies on input from the inner ear to sense linear and angular acceleration as well as to maintain a steady gaze and an upright vertical stance.

The skin on the bottom of the foot plays an important role in detecting balance information and is one of the most sensitive areas of the human body (Hosoda et al, 1998). Therefore since shoes alter the interface between these mechanoreceptors and the external environment, they must be taken into account when looking at factors that influence postural control. 45% of all falls can be attributed to improper footwear (Menant et al, 2008). Some of the footwear characteristics that have the potential to affect balance include sole thickness, sole hardness, and overall shoe mass (Robbins et al, 1994; Perry et al, 2007; Garner, Wade, Garten, Chander, & Acevedo, 2013). Fatigue has also been shown to have the ability to influence balance performance (Nardone, Tarantola, Giordano, & Schieppati, 1997).

While extensive research has been done to examine the effects of these shoe characteristics, new types of shoes have become more popular in recent years and differ from more traditional footwear in several ways. More research is needed to determine whether these new footwear styles affect balance differently than more traditional footwear. It has been observed that flip flops are associated with abnormal kinematics in the lower leg, but there is a lack of research on how they affect standing balance after walking (Shroyer, 2009). Because the clog-style Crocs have textured insoles, it is thought that they could possibly provide increased somatosensory feedback and therefore improved balance performance, but there is not adequate research to fully support that notion (Dixon et al., 2012; Hatton, Dixon, Martin, & Rome, 2009). Also, the minimalist style Vibrams have become popular recently because they are supposed to mimic the barefoot running experience while still providing just enough protection to prevent puncture wounds and cuts on the foot, but there is not yet sufficient scientific data to support these claims (Gangemi, 2011; Squadrone & Gallozzi, 2009).

Because footwear can have a significant adverse effect on balance (Gangemi, 2011), it is important to know which types provide the most stability. While many forms of footwear have been studied extensively, several of the new alternative styles lack any true scientific consensus. The purpose of this study was to observe balance measures such as sway velocity and sway root mean square (RMS) in the Crocs, flip flops, and Vibrams after a one mile walk to examine how these new footwear styles affect balance.

Purpose:

The purpose of this study was to determine the effects of alternative footwear on balance measures of sway velocity and sway RMS after a one mile walk at a self selected pace.

Hypotheses:

H_{O1}: There will be no difference in AP/ML sway velocity between the different types of footwear.

H_{A1}: There will be a significant difference in AP/ML sway velocity between the different types of footwear.

H_{O2}: There will be no difference in AP/ML sway RMS between the different types of footwear.

H_{A2}: There will be a significant difference in AP/ML sway RMS between the different types of footwear.

It has been shown in previous studies that differences in shoe characteristics such as sole thickness, hardness, and overall shoe mass can significantly impact balance measures (Robbins et al, 1994; Perry et al, 2007; Garner, Wade, Garten, Chander, & Acevedo, 2013). However, it is unclear exactly how much each variable contributes to overall stability. Also, it is unclear whether or not the variance seen among the three shoes in each of these characteristics is enough to cause significant differences in balance measures.

H_{O3}: There will be no difference in AP/ML sway velocity between pre and post measures of a one mile, preferred pace walk.

H_{A3}: There will be a significant difference in AP/ML sway velocity between pre and post measures of a one mile, preferred pace walk.

H_{O4}: There will be no difference in AP/ML sway RMS between pre and post measures of a one mile, preferred pace walk.

H_{A4}: There will be a significant difference in AP/ML sway RMS between pre and post measures of a one mile, preferred pace walk.

We have also seen that fatigue can have the potential to influence balance performance (Nardone, Tarantola, Giordano, & Schieppati, 1997). The one mile walk could possibly be expected to generate fatigue, thereby reducing balance performance in the post-walking values as compared to the pre-walking values. However, it is unclear as to whether or not the one mile walk will generate enough fatigue to make a difference in balance measures.

Definitions:

Balance: also known as postural control; a dynamic equilibrium between internal and external forces and environmental factors (Yaggie & McGregor, 2002). The maintenance of the center of gravity within the base of support (Winter et al, 1990).

Center of Mass (COM): the point on a body that moves in the same way that a particle subject to the same external forces would move (Rodgers, Cavanagh, 1984).

Center of Gravity (COG): the point at which the weight of the body or system can be considered to act and at which the weight of the body will be applied to create balance in relation to translational and rotational gravitational effects that act on the system (Rodgers, Cavanagh, 1984).

Center of Pressure (COP): describes the centroid of pressure distribution**;** sometimes referred to as the point at which the force is applied (Rodgers, Cavanagh, 1984).

Fatigue: a reduction in the ability of a muscle or muscle group to produce force (Decorte, Lafaix, Millet, Wuyam, & Verges, 2012)

Base of Support (BOS): the area defined in humans posteriorly as the tip of the heels and anteriorly as a line drawn between the tips of the toes; much smaller in humans than in quadrupedal species (Levangie & Norkin, 2011).

Perturbation: a variation of a system or process from its routine state; produced by an outside source (Winter, 1995).

Visual System: the system charged with gaining information about the environment and the body's position within it through the eyes (Sturnieks & Lord, 2008).

Somatosensory System: the system that involves tactile receptors and proprioception; consists of input from central nervous system touch receptors such as Ruffini endings, Merkel's disks, Pacinian corpuscles, and Meissner's corpuscles (Sturnieks & Lord, 2008).

Vestibular System: the system that obtains information about motion relative to body and eye movements, head posture and position, and gravity from structures in the inner ear (Sturnieks & Lord, 2008).

CHAPTER II

REVIEW OF LITERATURE

Balance, or postural control, is a dynamic equilibrium between internal and external forces and environmental factors (Yaggie & McGregor, 2002). The bipedal nature of human locomotion presents a unique challenge to the balance control systems of the body that maintain this equilibrium (Winter, 1995). Humans are inherently unstable, with two thirds of our body mass located two thirds of our body height above the ground, so these balance control systems must be working continuously and effectively (Winter, 1995). The visual, vestibular, and somatosensory or proprioceptive systems are the three major balance control systems. These systems of balance are responsible for determining the anterior/posterior and medial/lateral limits of stability (Yaggie & McGregor, 2002).

Vision is primarily used in planning locomotion and navigating obstacles (Winter, 1995). This visual input is used to make postural adjustments in anticipation of changes in surroundings (Nashner, 1982). However, under conditions of misleading or inappropriate visual input, the brain can suppress these signals in favor of vestibular or somatosensory input. For example, when exposed to linear or circular movements of visual surroundings, an individual may make anticipatory adjustments but will not lose balance (Nashner, 1982). Vision is also used while standing to monitor and moderate postural sway (Sturnieks & Lord, 2008).

The vestibular system aids balance by way of the vestibulo-ocular reflex which controls eye muscles and direction of gaze in response to movements of the head and changes of direction (Guskiewicz & Perrin, 1996). The organ responsible for detecting sensations of equilibrium is the vestibular apparatus, located within a chamber of the temporal bone known as the bony labyrinth (Guskiewicz & Perrin, 1996). Using input from the vestibular apparatus, the body can sense linear and angular acceleration, allowing the head and neck to remain in an upright position (Winter, 1995). The vestibular system is capable of balancing the body even under conditions of functionally inappropriate visual and somatosensory input (Nashner, 1982). Vestibular input, unlike visual and somatosensory input, is inertial and gravitational based and therefore cannot be affected by external context within an earthbound setting (Nashner, 1982). Research suggests that vestibular input is used primarily in stabilizing slow body sway, and that the body relies mostly on the visual and somatosensory systems to maintain balance under normal conditions (Guskiewicz & Perrin, 1996). However, when the head tilts or when the body comes under sudden perturbations, the vestibular system's ability to return the head to an upright position and the vestibulo-ocular reflex become crucial (Guskiewicz & Perrin, 1996).

The somatosensory system involves tactile senses such as touch, pressure, and vibration as well as the sense of position, or proprioception, which determines the relative location and movement of all body parts (Guskiewicz & Perrin, 1996). Meissner's corpuscles, Pacinian corpuscles, Merkel's disks, and Ruffini endings all supply the central nervous system with information about sensations of touch (Hijmans et al., 2007). Also, cutaneous mechanoreceptors in the feet provide tactile feedback that allows the CNS to determine how much pressure is being applied to each part of the foot, leading to a greater consciousness of the body's posture (Hijmans et al., 2007). Proprioceptors play a major role in the somatosensory system as well and include muscle spindles and Golgi tendon sensory receptors. Muscle spindles provide information about muscle length, while Golgi tendon receptors send information about muscle tension (Guskiewicz & Perrin, 1996). Myotatic reflexes use information from these proprioceptors to correct for changes in muscle length and maintain correct posture (Guskiewicz

& Perrin, 1996). Like visual input, somatosensory input can be suppressed if it is functionally inappropriate or misleading (Nashner, 1982).

When studying balance, it is important to differentiate between the body's centers of mass, gravity, and pressure. The point on a body that moves in the same way that a particle subject to the same external forces would move is known as the center of mass (COM) (Rodgers, Cavanagh, 1984). The location of the COM depends on the position of the body and may not necessarily be located inside the body (Rodgers, Cavanagh, 1984). The center of gravity (COG) is the vertical projection of the center of mass (COM) onto the ground (Winter, 1995). The COG is the point at which the weight of the body can be considered to act (Rodgers, Cavanagh 1984). The base of support (BOS), on the other hand, is the area where the body makes contact with the ground (Rodgers, Cavanagh 1984). If the COG is allowed to move outside of the BOS, a limb must move in order to compensate and keep the body from falling by expanding the BOS (Maki et al., 2008). Meanwhile, the center of pressure (COP) is completely independent of the COM. The COP is a weighted average of all pressures exerted onto the ground by the body (Winter, 1995). When standing, each foot has its own COP, but two separate force plates must be used to determine these individual points (Winter, 1995). The net COP, which lies in between the two feet, is often measured instead when only one force plate is available (Winter, 1995).

By using plantarflexors to control net ankle movement, the COP can be shifted to regulate the position of the COG (Winter, 1995). In the event of forward sway and an anterior shift of the COG, the body can activate plantarflexors to move the COP anterior to the COG. When the COP becomes anterior to the COG, angular acceleration will reverse, causing angular velocity to decrease until the body eventually moves in a posterior direction. Similarly, when the body senses that a posterior shift of the COG needs to be corrected, the COP is moved to a position posterior to the COG through reduced plantarflexor activation. In order for the body to maintain balance through ankle movement alone, however, the COP must have a greater dynamic range than the COG. For example, if the COG is allowed to move to the far anterior portion of the toes, the COP may not be able to move far enough anteriorly to reverse the anterior angular acceleration, and the body may have to move a limb forward to maintain balance (Winter, 1995).

The body has two main methods of maintaining balance without stepping forward. These methods are known as the ankle strategy and the hip strategy (Winter, 1995). The ankle strategy uses the plantarflexors and dorsiflexors of the ankle to control for minor perturbations (Winter, 1995). When the perturbations become stronger, the body can utilize the hip strategy to flex or extend the hip, moving the COM in a posterior or anterior direction respectively (Winter, 1995).

These control systems all work together to maintain balance under normal conditions, but there are several external factors that also influence balance. Fatigue, for example, can play a role in maintaining postural stability. Fatigue is a reduction in the ability of a muscle or muscle group to produce force (Decorte, Lafaix, Millet, Wuyam, & Verges, 2012). It has been found that fatiguing exercises, or those that cause the body to surpass the anaerobic threshold, can significantly increase sway path and especially sway area as compared to control levels (Nardone, Tarantola, Giordano, & Schieppati, 1997). However, this effect is most prominent when visual input is restricted, suggesting that visual input may correct for fatigued proprioceptors (Nardone, Tarantola, Giordano, & Schieppati, 1997). It is important to note, though, that these changes in sway were only transient and disappeared after fifteen minutes following the exercise (Nardone, Tarantola, Giordano, & Schieppati, 1997). In place of a generalized muscle fatigue resulting from aerobic exercise, localized fatigue of ankle plantarflexors has been shown to have a significant effect on balance as well (Corbeil, Blouin, Bégin, Nougier, & Teasdale, 2003). It appears that this localized fatigue impacts balance by affecting motor output of the postural control system, rather than by affecting sensory input (Corbeil, Blouin, Bégin, Nougier, & Teasdale, 2003).

Clearly, though, our external conditions must be taken into account when observing postural control. While some advertisers have claimed in the past that softer surfaces and mats can reduce fatigue, thereby increasing balance, little differences in COP patterns have been found when comparing hard and soft surfaces (Duarte, Harvey, & Zatsiorsky, 2000). However, when looking at dynamic, rather than static standing conditions, it has been shown that variance in surface conditions can cause specific and measurable changes in gait. Under wet surface conditions, for example, subjects were found to have a reduced walking velocity, decreased step length, and a flatter shoe-floor angle at heel strike (Menant, Steele, Menz, Munro, & Lord, 2009). Research suggests that subjects show differences in gait when walking on smoother surfaces as opposed to rougher surfaces but that the greatest differences in gait occur when subjects expect a contaminated surface (Cham and Redfern, 2002). These anticipatory changes result in a decrease in the required coefficient of friction (RCOF) needed to maintain balance, with the knees and hips being utilized more than the ankles to achieve these postural changes (Cham and Redfern, 2002). In another study by the same authors it was found that right after slipping, subjects tried to control for the slipping motion, sometimes even reversing heel motion, before the heel again slid forward resulting in a fall (Cham and Redfern, 2002). The authors found that slip distances greater than 14 cm and peak forward sliding velocities greater than 0.7-0.8 m/s invariably resulted in a total loss of balance and a fall (Cham and Redfern, 2002).

In addition to surface conditions, it is difficult to ignore the impact that footwear has on our stability and traction (Menant et al, 2008). The hairless skin on the bottom of the foot is one of the most sensitive areas of the human body, with mechanoreceptors continuously converting static and dynamic balance information into nerve impulses to be carried to the brain (Hosoda et al, 1998). Shoes can alter this somatosensory input and create a varying amount of friction between the floor and the shoe sole (Menant et al, 2008). Because of this effect on cutaneous and proprioceptive input, many advocates of barefoot running proclaim that running in little or no footwear can help reduce injuries (Rose, Bowser, McGrath, & Salerno, 2011). Some studies indicate an increased amount of dynamic balance while barefoot during a single leg jump landing test (Rose, Bowser, McGrath, & Salerno, 2011). This seems to suggest that shoes impede mechanoreceptor feedback. However, most scientists conclude that wearing appropriately fitted shoes is the best way to reduce falls, as opposed to wearing just socks or going barefoot (Menant et al, 2008).

Studies show that shoes with thin and hard soles tend to be the best for maximizing stability. While shoes with thick, soft soles are often selected for their comfort, these soft soles tend to increase ankle motion in the medial-lateral plane (Robbins et al, 1994). It is believed by some researchers that increased rapid ankle movements may lead to a poor sense of ankle proprioception and that soft soles may impact the body's ability to judge plantar pressure distribution (Robbins et al, 1994). Because they provide less of a support base, softer soles also make it harder for the body to counteract balance perturbations (Perry, Radtke, & Goodwin, 2007). Thus, as the softness of the sole of a shoe increases, the body must exert increasingly strong mechanical responses to maintain balance and becomes less able to deal with more severe perturbations (Perry Radtke, & Goodwin, 2007). In addition, when looking at soles, arch support and heel cups need to be taken into account. For example, when examining balance performance among varying styles of flip flops, one study found that flip flops with arch support and heel cups tended to lead to a more normal gait pattern than those without these features, suggesting that arch support and heel cups are beneficial to proper gait function (Shroyer, 2009). However, it is possible that arch support can lead to increased lateral sliding and a decreased stride length when compared to barefoot walking (Shroyer, 2009). Soles with stabilization and anti-pronation devices, as well as those with too much cushioning, have been shown to lead to unnatural gait patterns when compared to barefoot walking (Gangemi, 2011).

In addition to soles, heel height can also have an effect in determining balance. Studies have shown that high heels (9 cm) can cause the body's COP to shift anteriorly to the forefoot, even after just one hour of walking (Ko, Lee, 2013). Similarly, flat shoes (0.5 cm) can cause the COP to shift to the hindfoot (Ko, Lee, 2013). Heel height can induce excess plantar flexion or dorsiflexion in high and low heels respectively (Ko, Lee, 2013). This displacement of COP and overuse of specific muscles needed to maintain this increased plantar flexion or dorsiflexion can lead to musculoskeletal disorders (Ko, Lee, 2013). Therefore, medium heeled shoes (4 cm) are recommended, since they do not cause a displacement in COP (Ko, Lee, 2013). Indeed researchers seem to have found an acceptable range of heel heights and indicate that heels up to at least 4 cm seem to cause no problems with balance (Lindemann et al, 2003). In comparing heels of 1 cm, 2 cm, and 3.2 cm; no significant differences in balance was found (Lindemann et al, 2003). In fact one study found, in comparing shoes with heel heights of 2.1 cm, 3.5 cm, and 3.8 cm; that the shoes with heel heights of 3.5 cm and 3.8 cm performed better in balance testing than the shoe with a heel height of 2.1 cm (Chander, 2012). However, these results could be due to differences in shoe shaft height and sole surface area (Chander, 2012).

Shoe mass can even play a role in balance. In studying firefighter boot types, it was found that heavier types of boots led to a more pronounced decrease of maximum torque produced by the muscles of the ankles and knee (Garner, Wade, Garten, Chander, & Acevedo, 2013). Increased amounts of fatigue may be why heavier rubber boots were found to cause increased sway velocity as opposed to lighter leather boots (Garner, Wade, Garten, Chander, & Acevedo, 2013). Again though, differences in mass may be less important than other differences in shoe dimensions. For example, shaft height and sole surface area were seen to be more influential in determining balance performance than mass (Chander, 2012). When walking, shoe mass has been thought to play a role in stride length. Shoes with greater mass have been shown in some studies to correlate with longer stride lengths, possibly due to increased inertia (Mundermann, et al, 2003). However, when comparing different styles of flip flops, heavier flip flops do not always produce longer stride lengths. In one study, flip flops with greater mass were shown to have a shorter stride length than those with a reduced mass, suggesting that mass may not be the only factor affecting stride length (Shroyer, 2009).

Obviously, there are differences between alternative footwear such as flip flops and more traditional athletic footwear. However, flip flops are not the only alternative footwear that has been produced in recent years. Alternative styles such as Crocs® and Vibram® Five- Fingers have also entered the market and may possibly affect the biomechanics of the foot in different ways from more traditional shoes. The purpose of the present study is to see how these alternative footwear types affect standing balance and postural control.

CHAPTER III

METHODOLOGY

Purpose

The purpose of this study was to determine the effects of three different styles of alternative footwear on balance after a one mile walk at a self selected pace. The study focused on how the different styles of footwear affected human balance and postural control under quiet standing conditions. Participants were also tested for lower limb extremity muscle activity using EMG.

Participants

The participants in this study were 18 healthy, recreationally trained males between the ages of 18 and 44 years. All participants were required to fill out two forms prior to the experiment: the physical activity questionnaire (PAR-Q) and the seven day physical activity recall (7-day PAQ) in order to determine if they were physically active and healthy enough to participate. Written informed consent was obtained by the Institutional Review Board (IRB). Participant 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

Instrumentation

The NeuroCom EquiTest Balance Master® was used to assess balance control and postural stability under dynamic test conditions. The system uses an 18” x 18” dual forceplate to measure forces exerted by the feet and includes a moveable visual surround to affect visual input. The system uses a Motor Control Test (MCT) and a Sensory Organization Test (SOT) in measuring balance. For the MCT, the system uses unexpected anterior and posterior shifts, known as translations, of the forceplate. This displaces the COG and forces the body to shift to maintain balance. Latencies were defined as the time between the forceplate shift and the initiation of the muscle response in the legs. As part of the SOT, the forceplate and the visual surround can be “sway referenced,” meaning that the forceplate and visual surround can be made to follow the subject’s anteroposterior body sway. Sway referencing can effectively eliminate useful visual and somatosensory input, causing the central nervous system to rely on other senses. Four sensory conditions from the SOT were used: (1) standing with eyes open (EO) and (2) closed (EC) with the force plate fixed, (3) standing with the eyes open and the visual surround sway referenced (EOSRV), and (4) standing with the eyes open and the platform sway referenced (EOSRP). Latencies were defined as the time between the translation onset and the active response in the leg. COP data was collected from these tests and used to calculate AP/ML sway RMS and AP/ML sway velocity using the equations below:

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

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

Experimental Conditions

Participants were asked to be a part of three different experimental conditions. These included wearing thong style flip flops (FF), Crocs with clogs (CC), and Vibram Five-Fingers minimalist shoes (MIN). Sole hardness was measured on the Shore A hardness scale, which assigns a hardness value ranging from 0A (very soft) to 100A (hard/extra hard materials). Shoe measures are included in Table 2.

Table 2

	Crocs (CC)	Flip Flops (FF)	Vibrams (MIN)
Heel Height (cm)	2.0	1.4	0.4
Toe Height (cm)	1.6	1.4	0.3
Weight (individual shoe in oz)	7.0	4.0	6.0
Average Sole Hardness	51.2A	71.4A	99.0A
Midsole Hardness	68.4A	61.3A	66.63A

Experimental Procedures

A Repeated Measures, counter balanced design using within-subjects factor was used. Participants visited the Kevser Ermin Applied Physiology Laboratory and the Applied Biomechanics Laboratory at the University of Mississippi four times, with each time separated by at least 24 hours. The visits occurred as follows.

Day 1: Each participant was familiarized with the experiment and testing procedures after university approved informed consent was obtained. Measures such as height, weight, resting heart rate, and resting blood pressure were taken, and participants completed a physical activity readiness questionnaire (PAR-Q). If any participants were deemed unfit by answering "yes" to any question, they were asked to leave the study. After the questionnaire, the walking-gait trials protocol was explained, and participants were asked to walk on a 70 ft indoor track to determine a self selected pace.

Day 2: The first part of experimental testing began at the Applied Biomechanics Lab. Participants were prepared and randomized for one of the three footwear conditions. They were then asked to stand as still as possible on the NeuroCom plate for a Sensory Organization Test (SOT) and a Motor Control Test (MCT). Following the SOT and MCT, participants moved to the Kevser Ermin Applied Physiology Laboratory, and each one was evaluated while walking one mile on a treadmill at his or her self selected pace. This self selected pace was determined by walking 70 ft on an indoor track 6 times and taking the average of the pace times. The first and last 10 ft were not counted in determining mean pace of each trial. Each participant was allowed a short warm-up period at half of this selected pace prior to walking a mile on the treadmill. After collecting necessary data in the physiology lab, participants moved back to the biomechanics lab where the SOT and MCT were used again in the same manner as before to assess balance.

Day 3 and 4: The next two days followed the same protocol as Day 2 but with a different one of the three footwear conditions selected for each participant. After the fourth day, each participant had completed the procedure in each of the three types of footwear.

Statistical Analysis: With a predetermined alpha level of 0.05, results were analyzed in SPSS using a 2x3 repeated measures ANOVA [2 measurement times (pre/post walking) x 3 footwear types (FF, CC, MIN)] for each of the SOT conditions and for the MCT latency times.

CHAPTER IV

RESULTS

Analysis

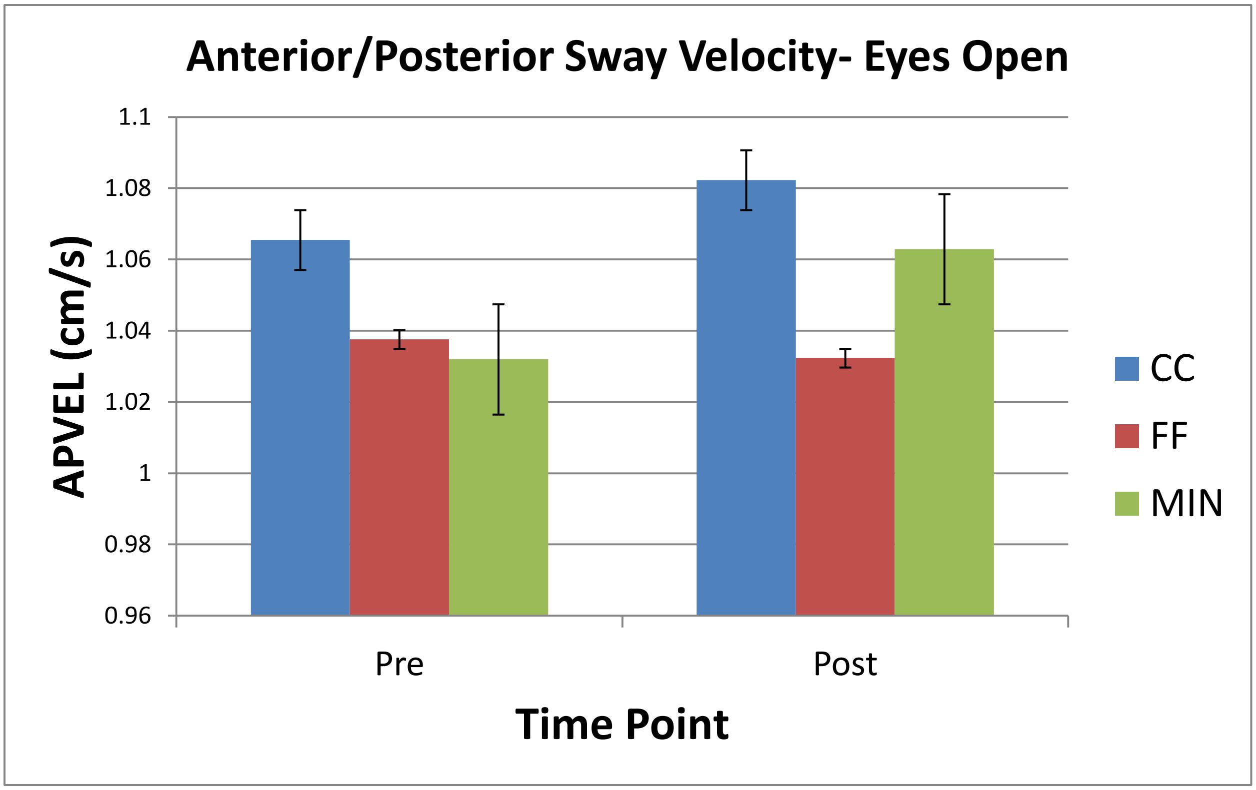
Static Balance was measured using the four conditions of the Neurocom® Equitest SOT (EO, EC, EOSRV, EOSRP). These conditions were numbered 1-4, with EO as condition 1, EC as 2, EOSRV as 3, and EOSRP as 4. The values of the average sway velocity and root mean square of the COP were used to measure postural sway in the anterior-posterior direction (APVEL and APRMS) and the medial-lateral direction (MLVEL and MLRMS). A repeated measures analysis of variance (ANOVA) was used to examine these values across the four SOT conditions. Values were also compared across two time points (pre-walk and post-walk). Significance was set at an alpha level of $p \leq 0.05$. If a significant main effect or interaction effect was found, a pairwise comparison with a Bonferroni correction was used to identify post-hoc differences.

Anterior-Posterior Sway Velocity

A significant difference was found when looking at footwear main effect ($p = 0.002$, $\eta^2 = 0.305$) on APVEL under condition 3 (EOSRV). In the pairwise comparison for condition 3, significant differences were found between the CC and the MIN ($p = 0.014$) and the FF and MIN ($p = 0.010$). No significant effects were found for footwear-time interaction, so the differences in sway velocity values can be generalized across the two time points. There were no significant effects seen for footwear, time, or footwear-time interaction under conditions 1, 2, and 4 for APVEL.

Figures: Averaged Sway Velocity measures in the Anterior-Posterior direction for each of the four 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.

Figure 1



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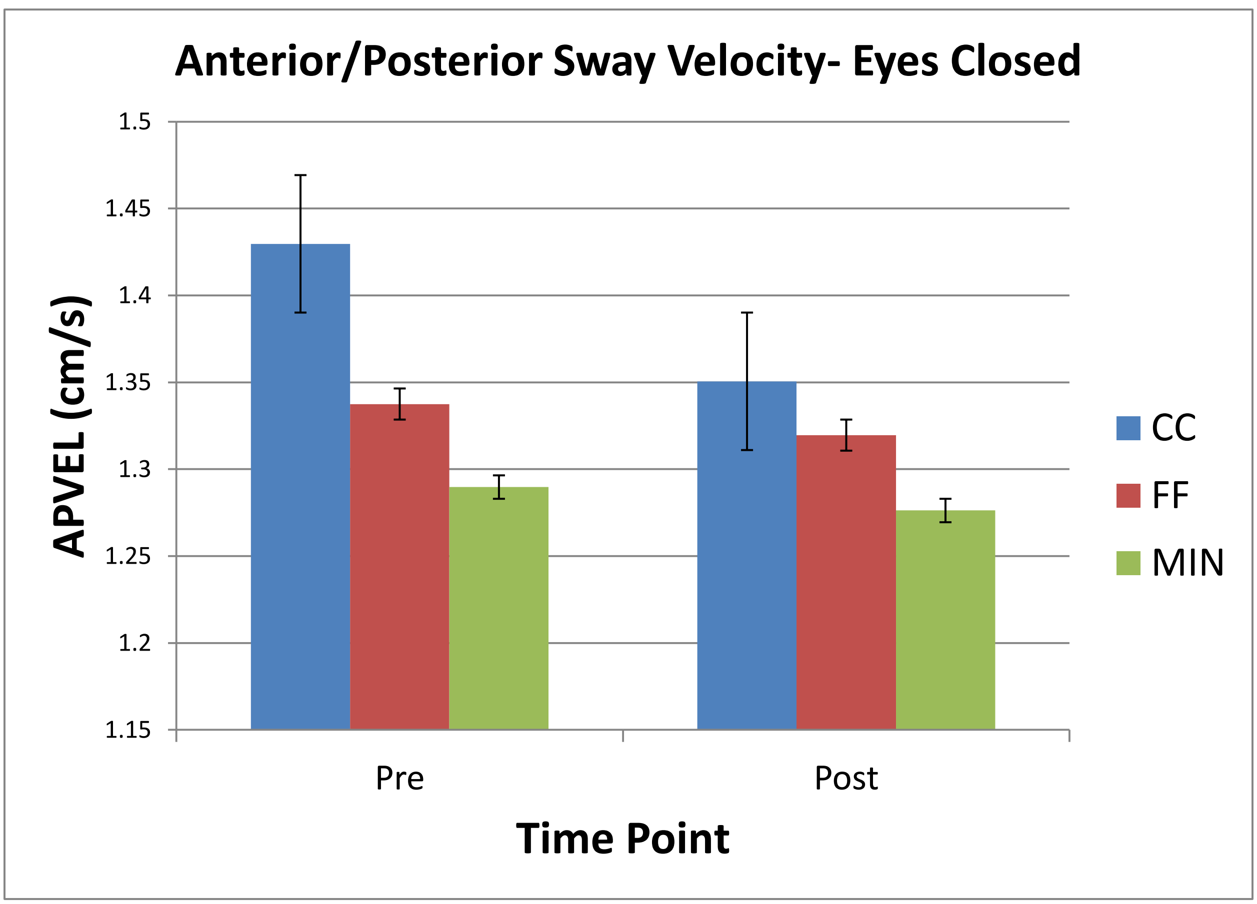

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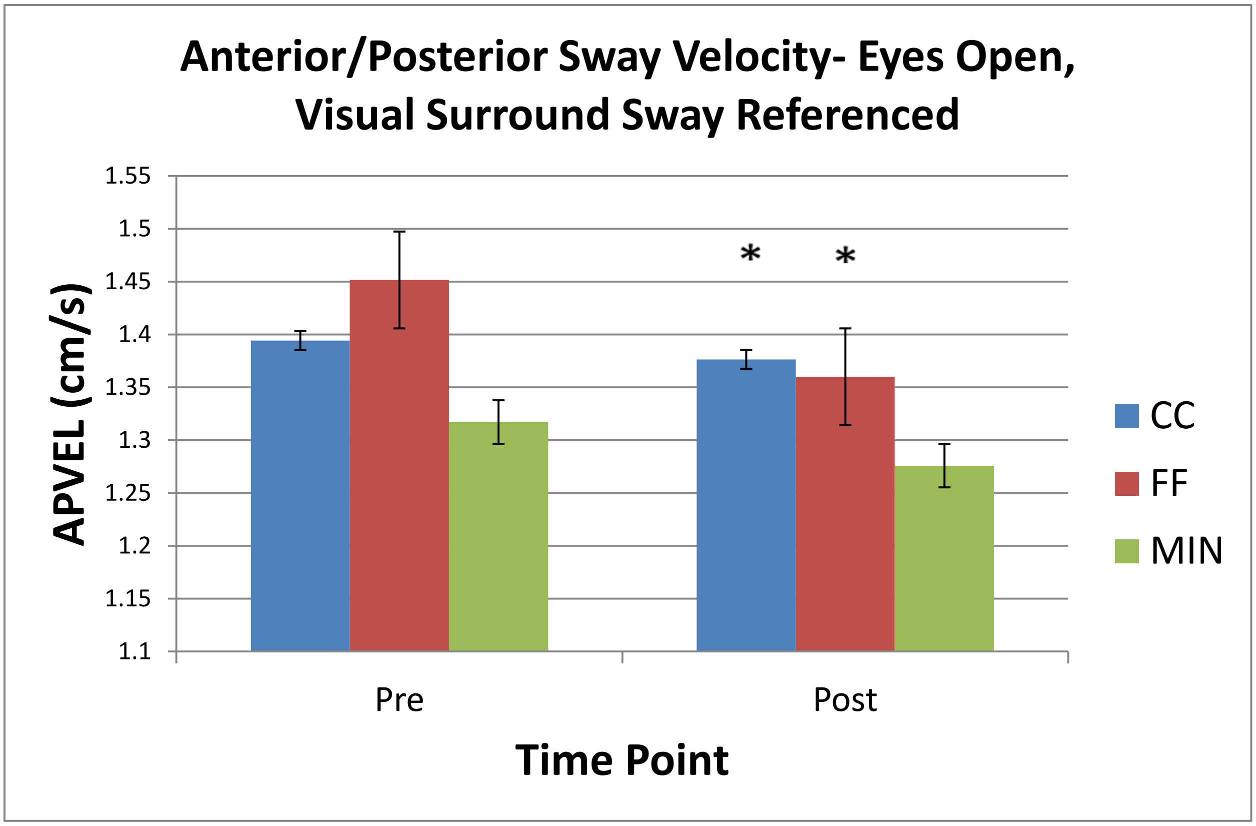

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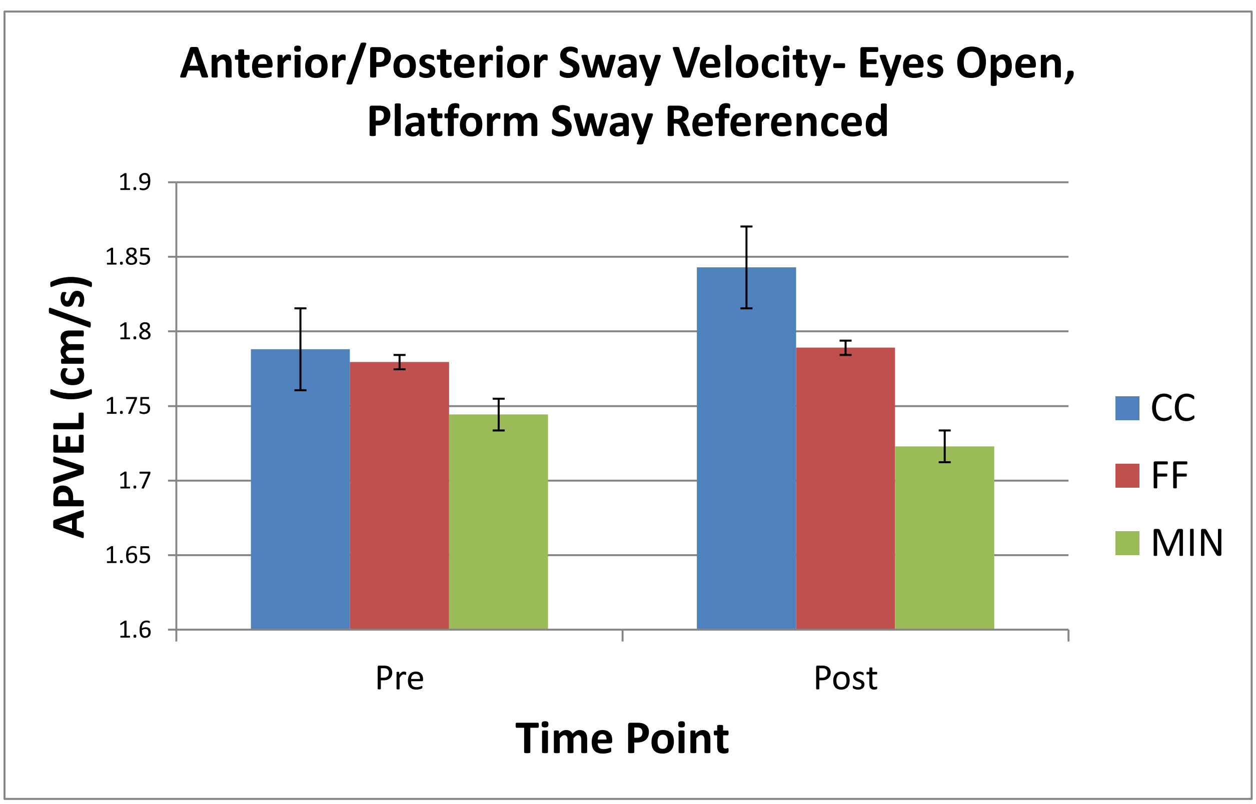


Anterior-Posterior Sway RMS

There was a significant difference in footwear main effect ($p=0.002$, $\eta^2= 0.318$) under condition 2 (EC) for APRMS. In the pairwise comparison for condition 2, a significant difference in APRMS was found between the CC and the MIN ($p= 0.001$). Because there was no significant effect by footwear-time interaction, the footwear effects can be generalized across the time points. There were no significant differences in APRMS found for footwear main effects, time main effects, or footwear-time interaction effects for condition 1 and conditions 3 and 4.

Figures: Averaged Sway RMS measures in the Anterior-Posterior direction for each of the four 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.

Figure 5

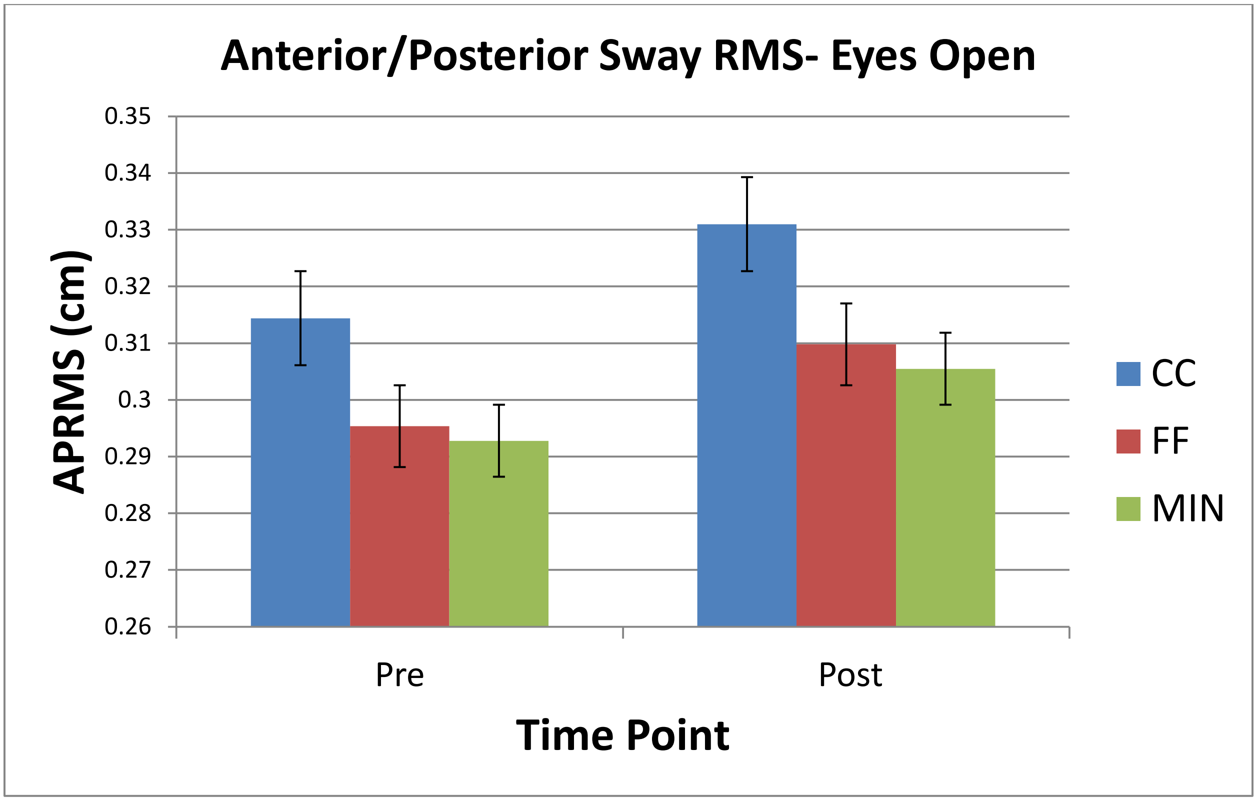


Figure 6

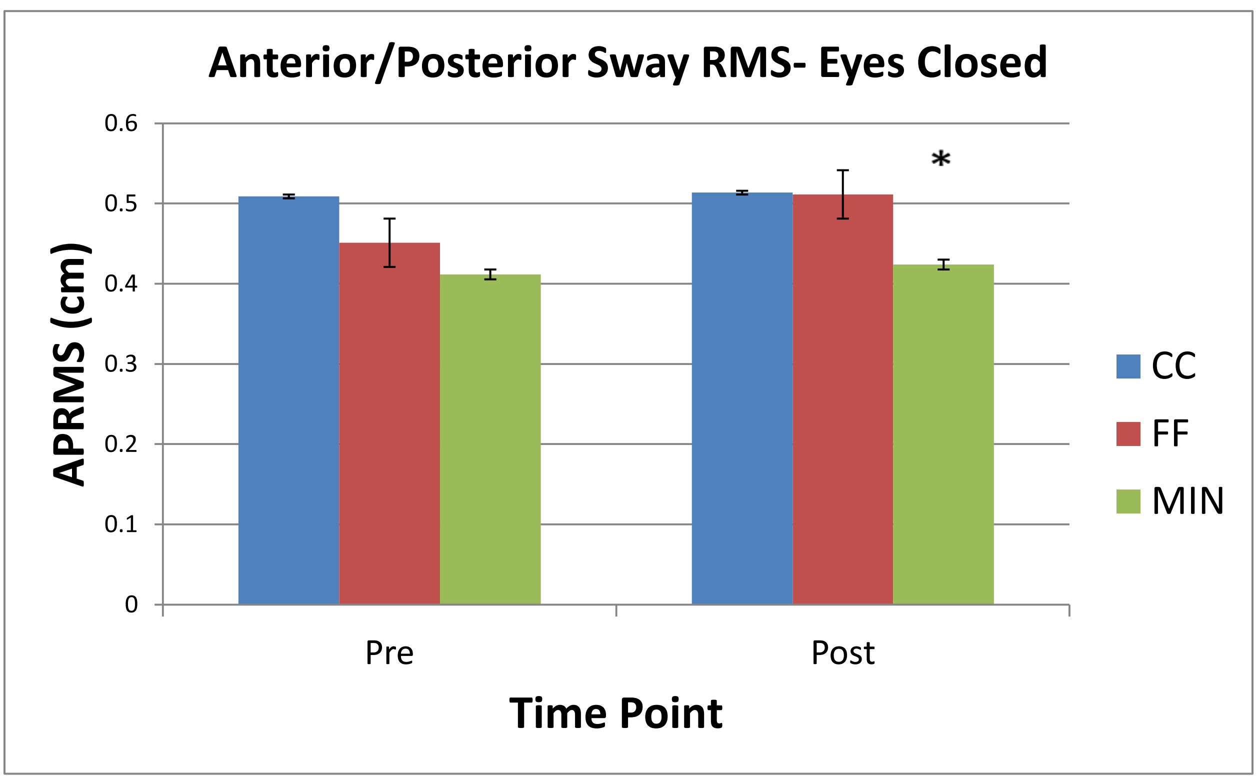


Figure 7

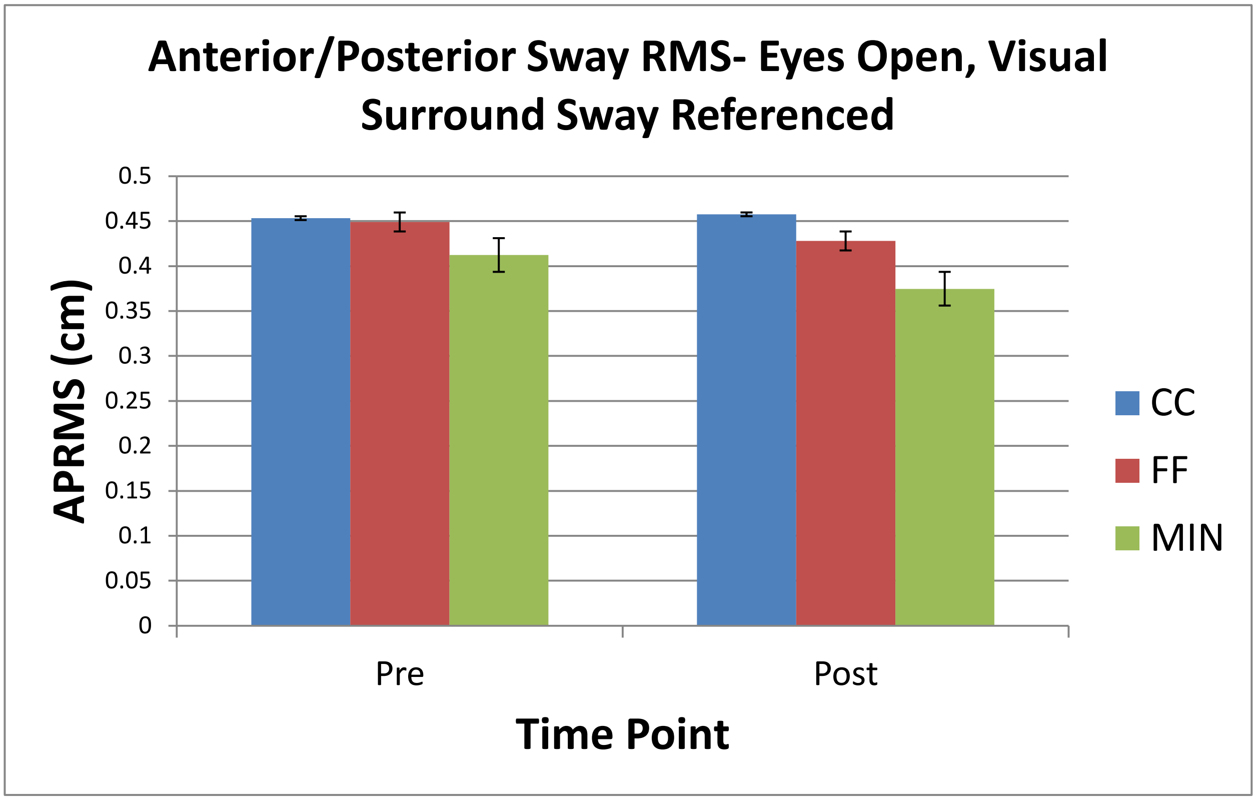
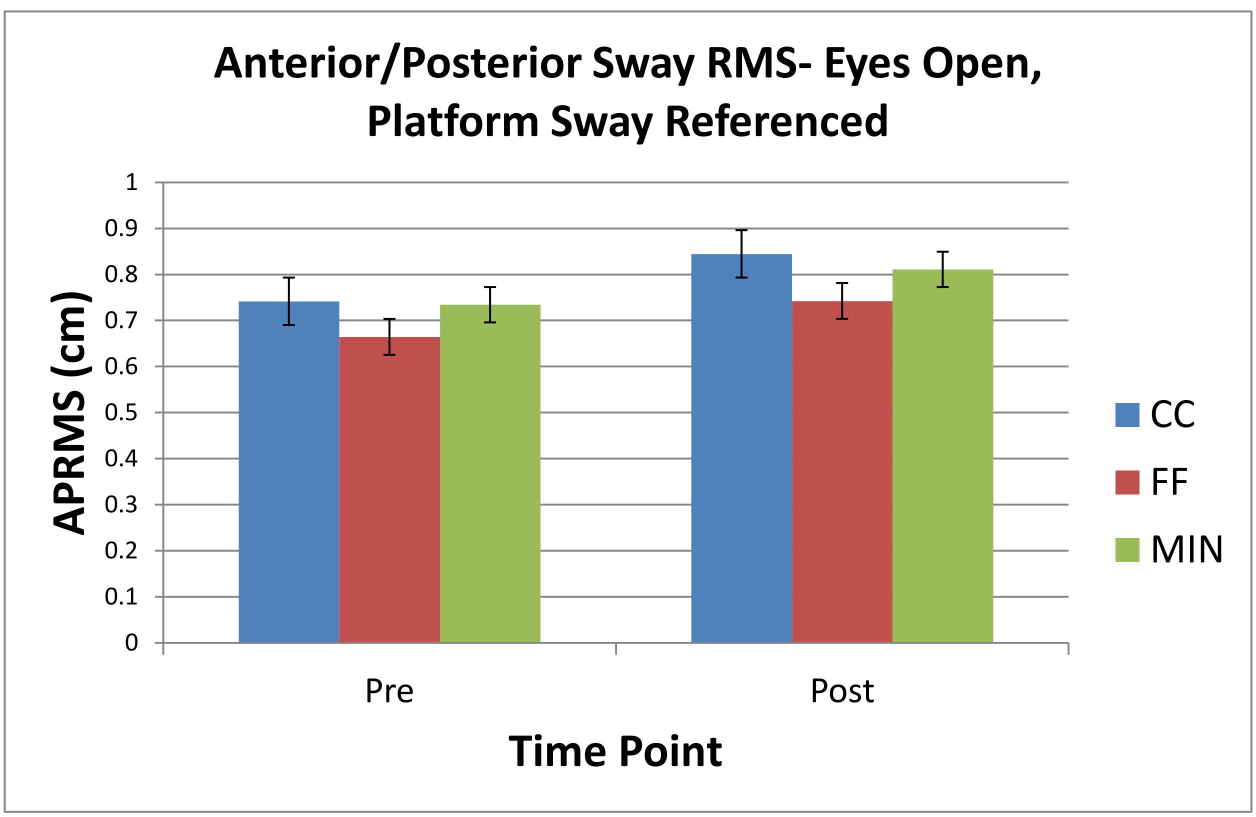


Figure 8



Medial-Lateral Sway Velocity

A significant difference was found in footwear main effect ($p=0.036$, $\eta^2= 0.178$) on MLVEL under condition 3 (EOSRV). In the pairwise comparison for condition 3, a significant difference was seen between the FF and the MIN ($p= 0.010$). No significant interaction effects were seen for condition 3, so the footwear effects can be generalized across the two time points. There were no significant effects seen for footwear, time, or footwear-time interaction for MLVEL under conditions 1, 2, and 4.

Figures: Averaged Sway Velocity measures in the Medial-Lateral direction for each of the four 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.

Figure 9

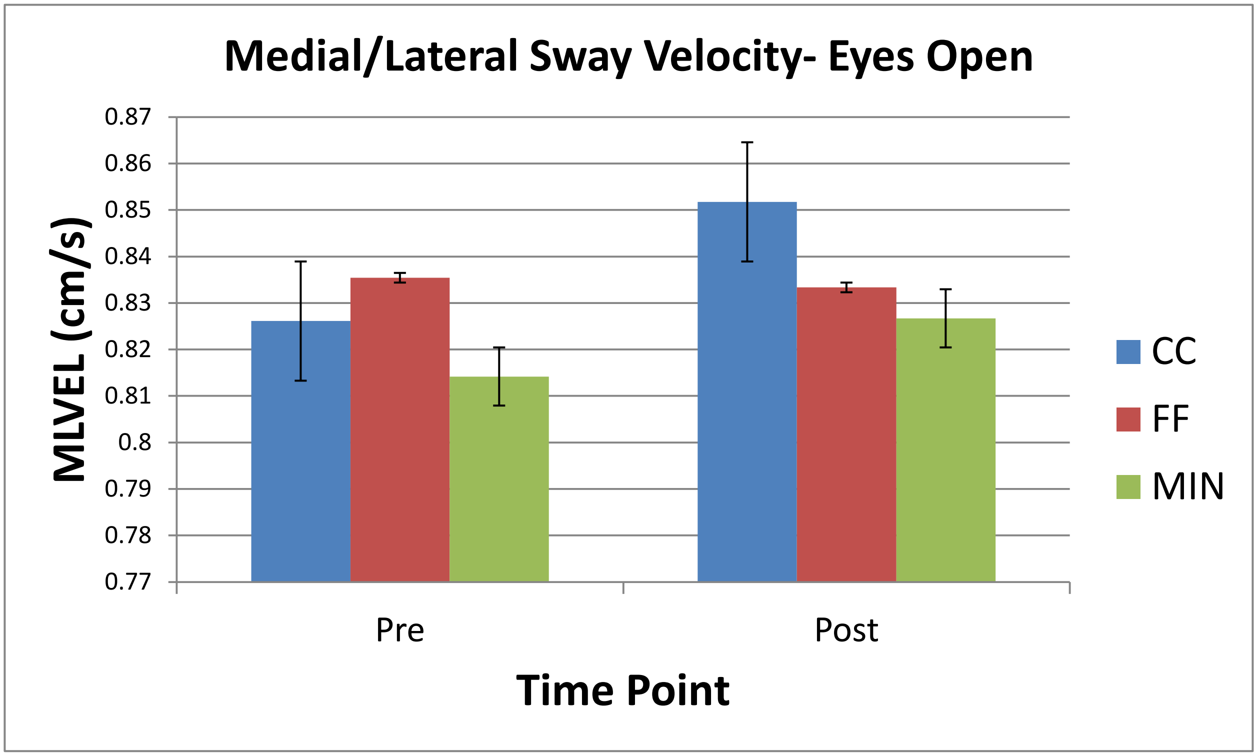


Figure 10

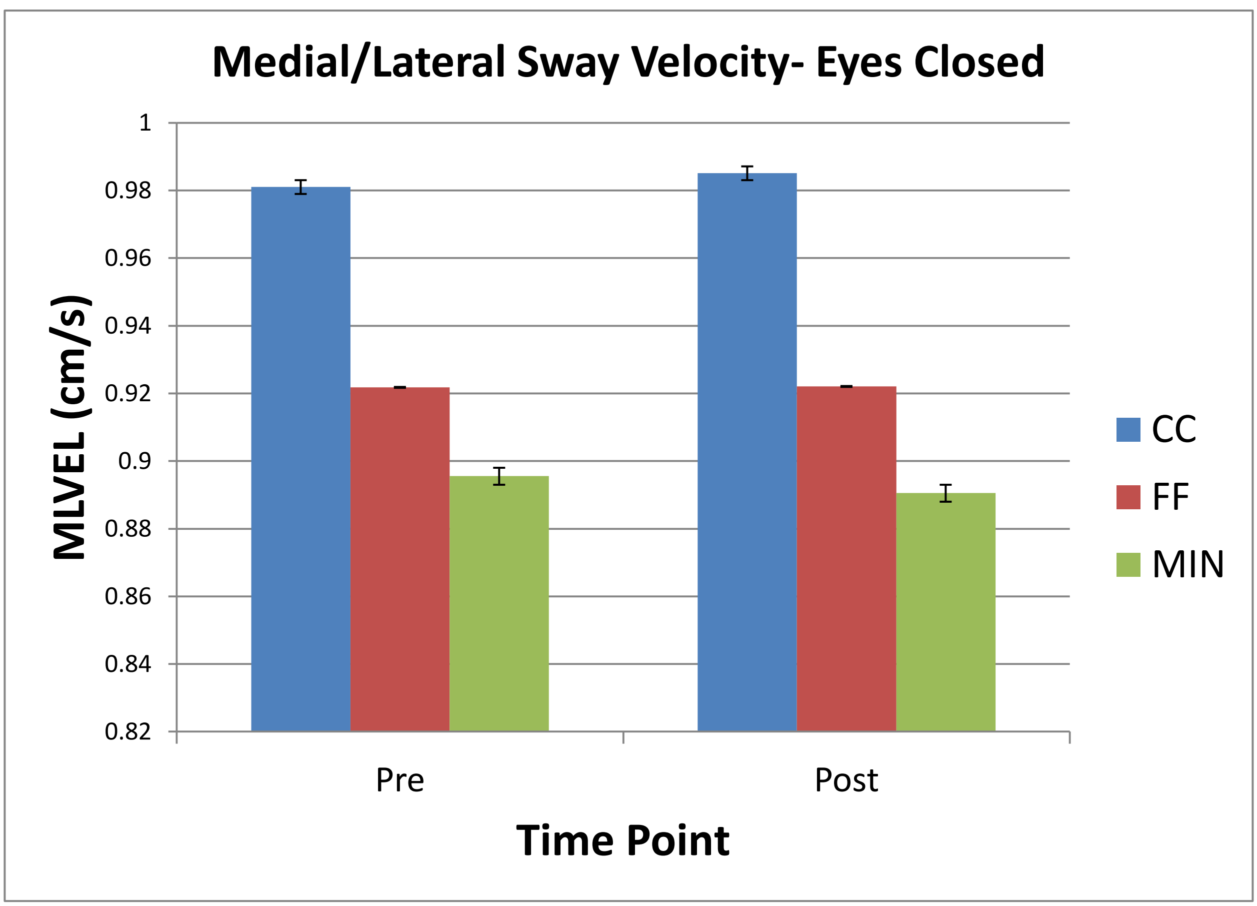


Figure 11

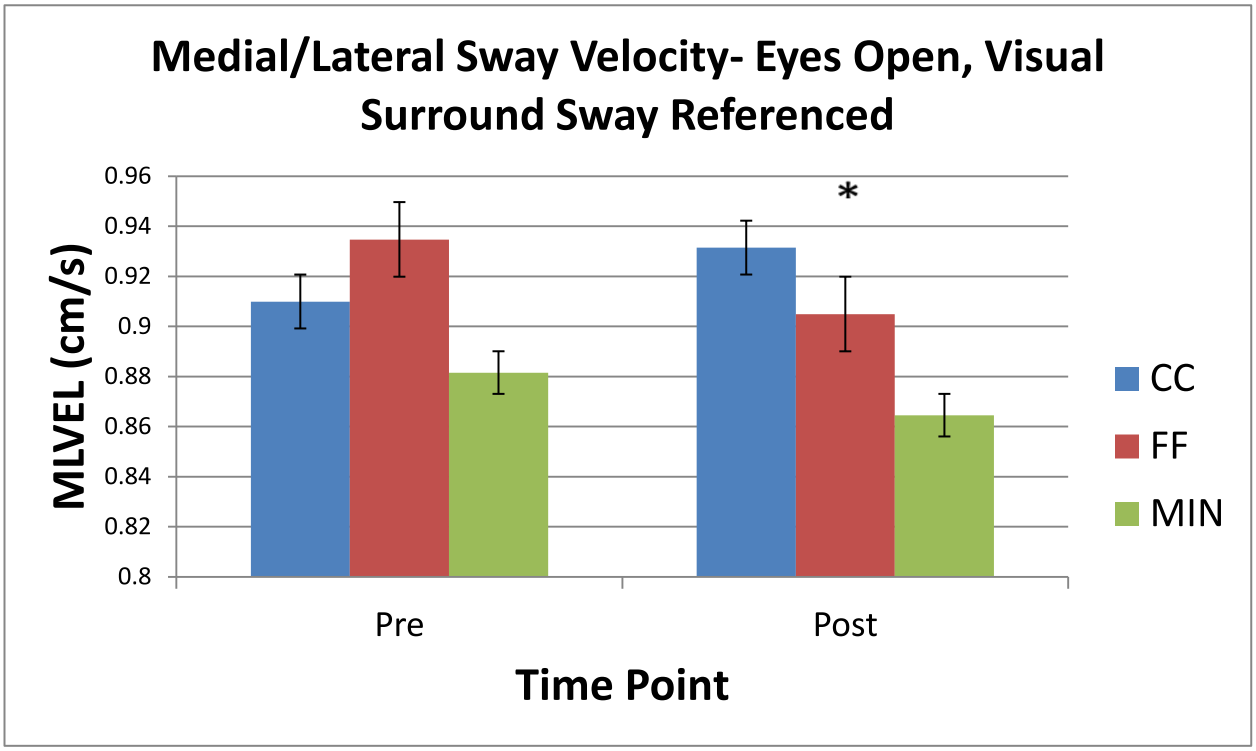
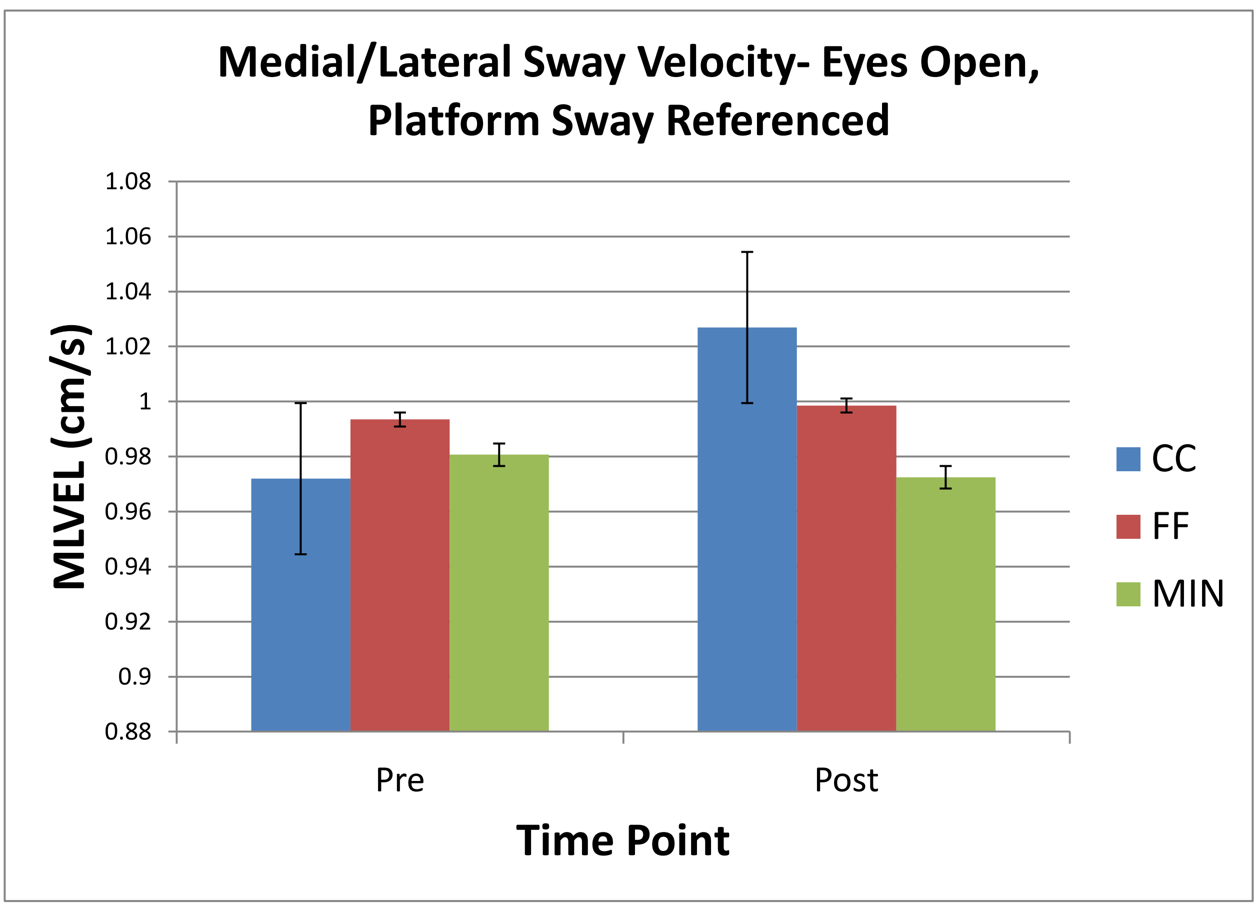


Figure 12



Medial-Lateral Sway RMS

Significant differences were found under condition 3 (EOSRV) for time main effect ($p= 0.020$, $\eta^2= 0.278$) and under condition 4 (EOSRP) for time main effect ($p= 0.008$, $\eta^2=0.346$). No significant effects were found for footwear-time interaction under condition 3 or 4, so the effects of time can be generalized across footwear types for both conditions. There were no significant effects seen for footwear, time, or footwear-time interaction for MLRMS under conditions 1 and 2.

Figures: Averaged Sway RMS measures in the Medial-Lateral direction for each of the four 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.

Figure 13

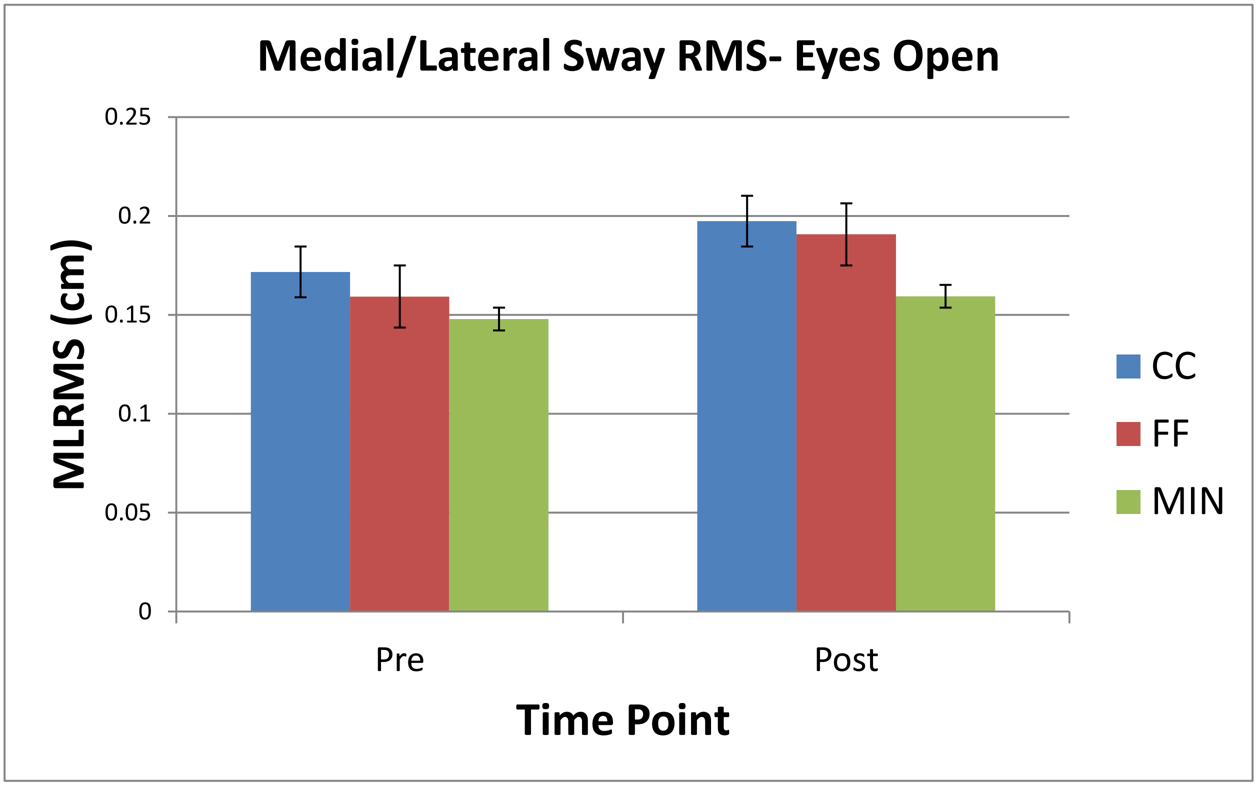


Figure 14

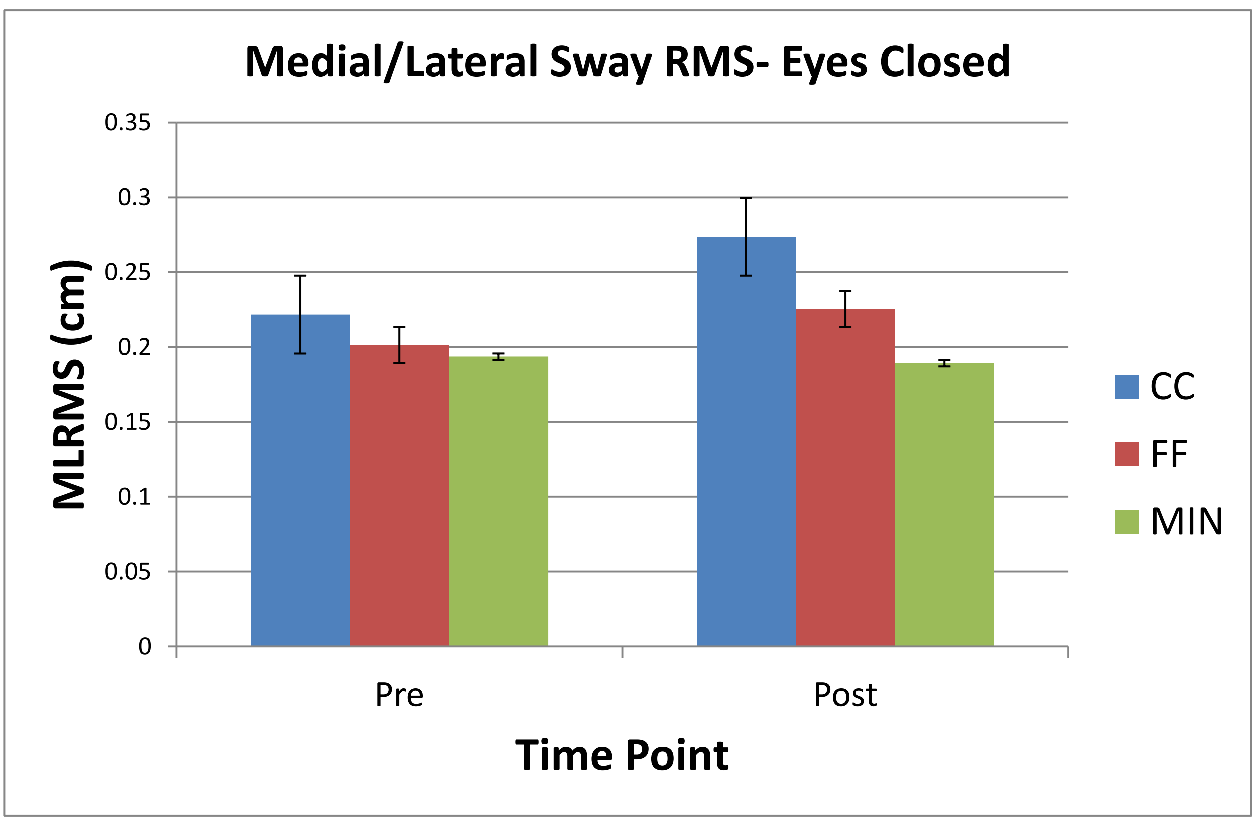


Figure 15

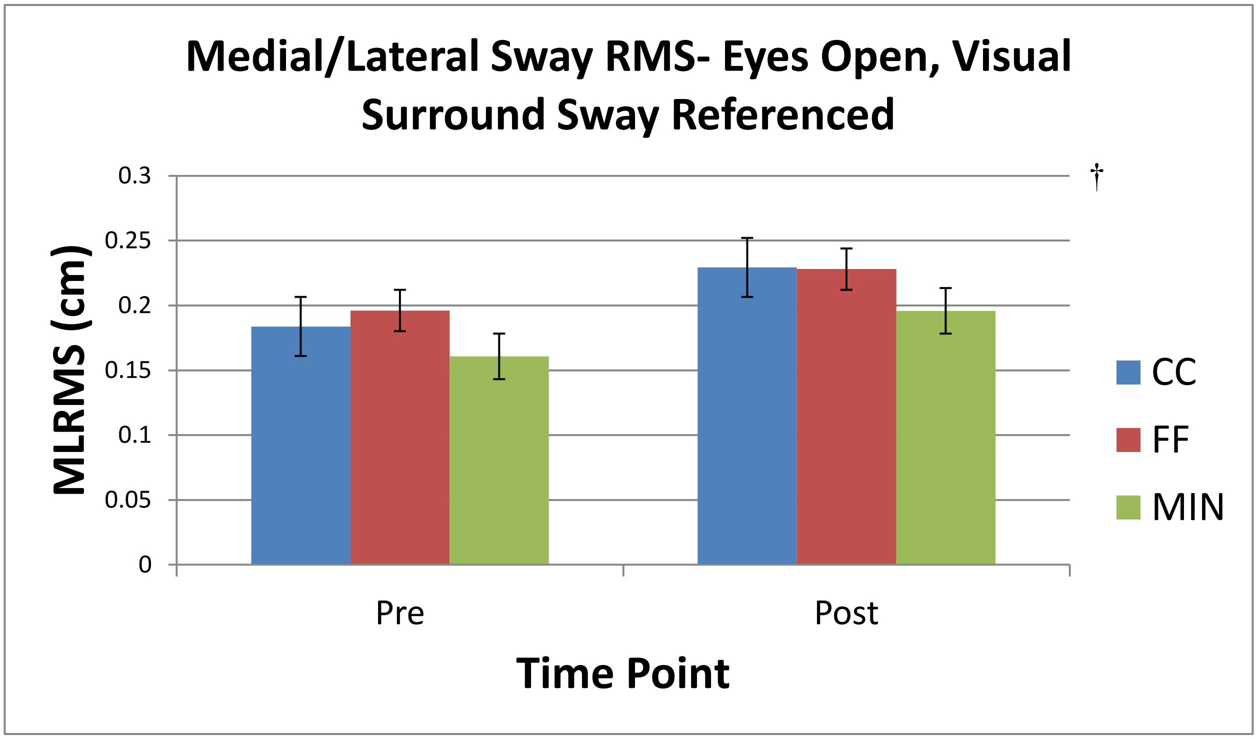
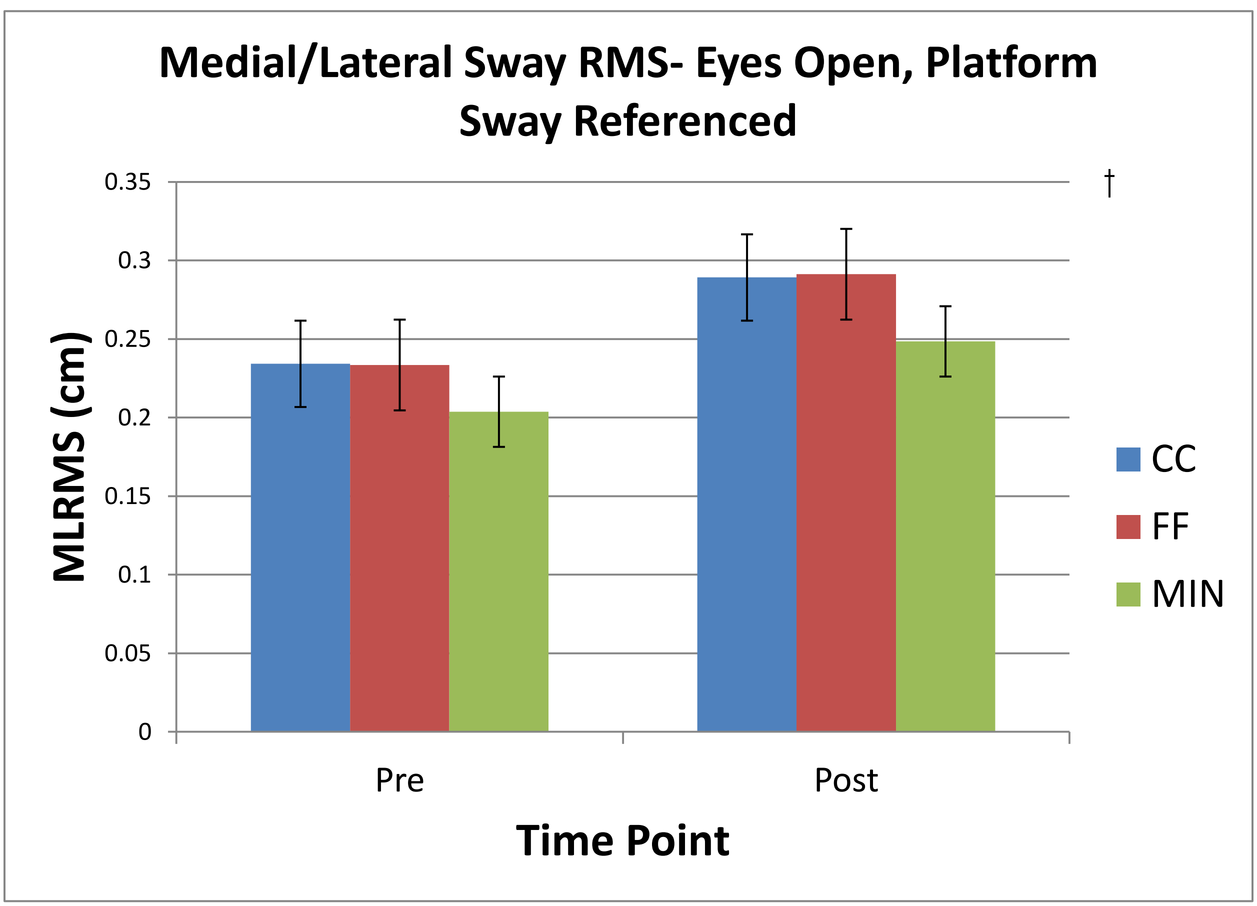


Figure 16



CHAPTER V

DISCUSSION

The purpose of this study was to determine whether or not alternative styles of footwear have an impact on postural control after a one mile walk. The three footwear styles were the Crocs (CC), flip flops (FF), and Vibrams Five-Fingers minimalist shoes (MIN). By using four conditions of the SOT (EO, EC, EOSRV, EOSRP) to observe each of the four postural control measures (APVEL, APRMS, MLVEL, MLRMS), variance was observed across footwear type. Significant differences were found between the CC and the MIN for the measures of APVEL and APRMS and between the FF and the MIN for the measures of APVEL and MLVEL. There was a significant main time effect for only one of the four postural control measures (MLRMS).

One possible reason for the difference in performance across footwear could be sole thickness. The MIN had the thinnest soles of the three footwear types (0.4 cm heel height and 0.3 cm toe height). The FF and CC were both at least 1 cm thicker than the MIN at the heel and toe. Also, the MIN had the hardest soles of the three shoes, with an average sole hardness of 99.0A as compared to the FF and CC at 71.4A and 51.2A respectively. Sole hardness was measured on hardness scale Shore A, which assigns a hardness value ranging from 0A (softer than a rubber band) to 100A (harder than a shopping cart wheel). These results are consistent with previous studies which found that hard and thin soles tend to outperform soft and thick soles in measures of balance (Robbins et al, 1994). These previous studies have shown that soft, thick soles tend to increase ankle motion in the medial-lateral plane (Robbins et al, 1994). However, Menant et al found that soft, thick soles did not really have an effect on standing balance but rather on stability during walking (Menant et al, 2008). Future studies could look into the effects of these three shoes on more dynamic measures of balance.

Another factor that has been shown in previous studies to have an effect on balance is shoe mass. Some studies have found that heavier shoes induce increased amounts of sway velocity due to greater amounts of fatigue (Garner, Wade, Garten, Chander, & Acevedo, 2013). Other studies have shown that mass may not be as important as other factors in determining balance performance (Chander, 2012). In the current study, the FF was the lightest shoe of the three at 4 oz. The MIN and the CC were more similar in mass at 6 oz and 7 oz respectively. Despite its lighter mass, the FF did have a significantly greater sway velocity in both the anterior-posterior direction and the medial-lateral direction than the MIN. This suggests that, under the conditions of this study, another factor may be more important in determining balance than mass. One reason for this could be that in the current study, subjects were only required to walk one mile on a treadmill. More fatiguing exercises may show a greater effect due to mass. Also, the difference in mass between the FF and the MIN (2.0 oz) may not be great enough to show any real effect due to mass. Future research could focus on determining whether or not there is a threshold shoe mass value at which fatigue becomes a factor that affects balance performance.

Despite textured insoles, the CC did not exhibit significantly better performance than the other two shoes in any of the balance measures or conditions. This seems to support earlier findings that textured insoles do not have a significant impact on balance performance (Hatton, Dixon, Martin, & Rome, 2007). However, because there were other significant differences between the three shoes besides textured insoles, future research could isolate textured insoles as a possible cause of changes in balance. Hatton, Dixon, Martin, & Rome isolated textured insoles as a variable, but there were only 8 participants in the study, and a larger sample size would be beneficial. While flip flops have been found to lead to decreased dorsiflexion when walking

(Shroyer, 2009), this change in kinematics did not seem to impact standing balance measures. The FF did not perform any worse than the CC during any of the studies, since both performed significantly worse than the MIN in two measures of balance apiece. Although, future research could compare the flip flop with a standard running shoe of similar mass and sole thickness to determine if the thong-style build of the shoe is responsible for differences in balance performance.

A significant main effect was seen for time for the measure of MLRMS, but since it only appeared as a main effect for one measure, it does not seem that time is a major contributing factor to balance performance under the conditions of this experiment. The one mile walk may not be strenuous or long enough to elicit enough fatigue to make any difference in postural control. Also, previous studies have found that changes in sway due to fatigue are typically transient and often disappear within fifteen minutes of the exercise (Nardone, Tarantola, Giordano, & Schieppati, 1997). The latency time between the walk and the balance testing may have been long enough for some fatigue effects to wear off. Future studies could minimize this latency time and also observe the effects of localized fatigue on these footwear types instead of just studying general aerobic fatigue caused by walking, since localized fatigue has led to impacts in postural control in some previous studies (Corbeil, Blouin, Bégin, Nougier, & Teasdale, 2003).

Conclusion

In this study, we see that there are significant differences in APVEL and APRMS between the Crocs (CC) and Vibrams (MIN). There were also significant differences in APVEL and MLVEL between the flip flops and Vibrams (MIN). There was only one balance measure in which time played a significant role in affecting balance performance (MLRMS). These results suggest that the decreased sole thickness and increased sole hardness of the MIN as compared to the CC and FF may have played a role in its better balance performance. Fatigue induced during the one mile walk may have had a small effect on balance performance, as indicated by the differences in MLRMS over time, but it did not have as strong of an effect as footwear. Based on the results of this study, future alternative footwear designs should focus on decreasing sole thickness and increasing sole hardness.

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