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Impacts on Balance When Walking in Occupational Footwear

Harish Chander

IMPACTS ON BALANCE WHEN WALKING IN OCCUPATIONAL FOOTWEAR

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HARISH CHANDER

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ABSTRACT

Hazards and challenges present in the workplace pose a number of potential risks for injuries and illness. Nearly 3.1 million nonfatal workplace injuries and illness were reported in 2010 (BLS, 2010). The probability of falls has been related to balance decrements. Further, an important point of distinction is 45% of all falls have been attributed to inappropriate footwear (Menant et al. 2008) Previous studies have shown decrements in balance as a result of different footwear (Menant et al. 2008) and after an increased workload over a specific period of time (Yaggie & McGregor, 2002; Gribble & Hertel, 2004). Occupational footwear is often designed for safety and may fail to provide appropriate foot biomechanics. As such the functionality of occupational footwear may impact balance characteristics over time. The purpose of the study is to examine the differences in balance in while wearing different types of occupational footwear for extended durations. Fourteen healthy male adults (aged 23.6±1.2 years; height of 181±5.3 cm; weight of 89.2±14.6 kg), with no history of orthopedic, musculoskeletal, cardiovascular, neurological and vestibular abnormalities participated in this study. The experimental session included an extended duration of walking (4hours) with balance measured at 30min intervals (Pre, 30, 60, 90, 120, 150, 180, 210 & 240min). The standing balance protocol assessment was done on the six conditions of the Neurocom Equitest SOT (EO, EC, EOSRV, EOSRP, ECSRP and EOSRVP). The values of the dependent sway variables were derived from the Center of Pressure (CoP) movement. The average sway velocity (VEL) and the root-mean-square (RMS) of the CoP were used to characterize the postural sway in the anterior-posterior (APVEL & APRMS) and the medio-lateral (MLVEL & MLRMS) directions during the 60-second testing period. Participants were randomly assigned 3 different types of occupational footwear: Work Boots (WB) (mass 0.39±0.06 kg), Tactical Boots (TB) (mass 0.53±0.08 kg) and Low Top Boots (LT) (mass 0.89±0.05 kg) with a minimum of 72 hours of rest between conditions. Balance dependent variables were evaluated using a 3 x 9 (Footwear [WB v. TB v. LT]) x (Extended duration of walking intervals [Pre, 30, 60, 90, 120, 150, 180, 210 & 240] RMANOVA and independently for the six SOT balance conditions (EO, EC, EOSRV, EOSRP, ECSRP and EOSRVP) to identify any existing differences within the exposure time as well as the footwear types. Significant differences were found over time in the EO, EC, EOSRV & EOSRP for MLRMS and between footwear in the EC for APRMS and MLRMS and EOSRP for MLRMS. These results indicate a decrement in balance performance over time but the differences were limited to MLRMS. The decline in balance may be attributed to fatigue resulting from an extended duration of walking/standing. Significant differences were found between the WB, TB and LT, where the LT had a higher postural sway RMS. The use of LT resulted in a relatively greater balance decrement, especially when vision was absent and with conflicting somatosensory input. The WB and TB despite having a greater mass, had less balance decrement, which can be related to their elevated boot shaft height. Results from this data suggest that the high boot shaft supports the ankle, resulting in decreased fatigue, and thus better balance.

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LIST OF ABBREVIATIONS

CoP – Center of Pressure

CoG – Center of Gravity

CoM – Center of Mass

LoG – Line of Gravity

BoS – Base of Support

AP – Anterior-Posterior direction

ML – Medio-Lateral direction

VEL – Sway Velocity

RMS – Root Mean Square

GRS – Global Reference System

SOT – Sensory Organization Test

TTB – Time To Boundary

CHAPTER I

INTRODUCTION

Nonfatal workplace injuries and illness among private industry employers declined in 2009 to a rate of 3.6 cases per 100 equivalent full time workers, down from 3.9 cases in 2008 as stated by the Bureau of Labor Statistics. The number of nonfatal occupational injuries and illness reported in 2009 declined to 3.3 million cases, compared to 3.7 million cases in 2008 from the estimates of the Survey of Occupational Injuries and Illness (SOII) published using the North American Industry Classification System (NAICS). However, the incidence rate for occupational injury and illness has gone up in the heavy and civil engineering construction and fire protection engineering from 12.9% to 13.1% and 14.8% to 15.3% respectively from 2008 to 2009 as reported by the Injuries, Illness, and Fatalities Program (IIF) by the Bureau of Labor Statistics.

Proper postural control and balance are essential in industrial settings in order to prevent falls and, thus, injuries. Increased probability of falls have been related to decrements in balance control and these falls are often a primary causative factor for injuries and disabilities in the general population as well as in the contemporary industrial population where postural stability is challenged with unfavorable and unfamiliar environment (Lin, Seol, Nussbaum & Madigan, 2008). In an occupational setting, postural instability can be hazardous due to increased risk of falls, slips, trips and other accidents (Kincl et al, 2002). Slips, trips and an induced loss of balance have often been reported as the primary causes of occupational falls (Redfern & DiPasquale, 1997; Maki et al., 2008).

Postural control is regarded as a skill where the central nervous system learns using information from passive biomechanical elements, sensory systems and the muscles. The postural control in the standing position works to control the body's orientation in space, to maintain the body's center of mass (CoM) over the base of support (BoS) (Winter, 1995; Levangie & Norkin, 2006) and to stabilize the head with regard to the vertical for the appropriately oriented eye gaze. A higher level of cognitive neural function and an intact neuro-muscular system is required for maintaining a normal posture/ balance and gait (Levangie & Norkin, 2006). The central nervous system interprets and organizes inputs from various structures and systems such as the visual system, somatosensory (proprioceptive) system and the vestibular system. The postural control mechanism is a result of the central nervous system, visual system, vestibular system and the musculo-skeletal system all working together in perfect harmony. In addition to the integrity of these systems, it also relies on information received from the receptors such as the joint receptors, golgi tendon organ, muscle spindle, proprioceptors, cutaneous and sensory receptors located in and around the joints and sensory receptors on the sole of the feet (Levangie & Norkin, 2006). Although these three major systems are responsible for maintaining balance, degradation or defect in any one of the system increases the probability of lowering balance performance and hence a possibility of a fall (Lepers et al. 1997).

The effects of extended durations of walking / standing on balance:

Standing and walking for long periods has been reported to cause a number of health related problems, particularly in the lower extremity (Cham & Redfern, 2001). These problems are particularly prevalent among workers who stand and walk for long periods of time. The extended durations for work hours place an increased workload over a prolonged time period leading to potential fatigue. All these deleterious effects of prolonged standing and walking are more pronounced with the lack of appropriate footwear. Among the many physical hazards and stresses that are common in the workers population are: lifting and carrying heavy loads; working in frequent awkward postures; walking with high risk of slips / trips / falls on irregular terrain; risk of accidents caused by the sudden unpredictable actions; and exposure to whole body vibrations. There are a number of contributing factors to biomechanical deficits in balance and posture and different occupational work settings along with the inappropriate foot and personal equipment may lead to musculoskeletal disorders such as osteoarthritis, rheumatoid arthritis, low back pain, upper limb disorders and hand-arm vibration syndrome, as well as consequences of trauma such as sprains, fractures and dislocations.

Although only a relatively small amount of muscular activity is required to maintain a stable erect standing posture, the control of posture is complex and involves the motor control system (Gefen, Megido-Ravid, Itzchak & Arcan, 2001; Caron, 2003). However, the activity of the musculature varies depending upon the different static and dynamic postural stability and balance measures. This ultimate goal of maintaining posture and balance in static and dynamic conditions are put in jeopardy when the much needed activity of the musculature is experiencing fatigue. Muscular fatigue can be defined as an inability of the muscle to maintain a reasonably expected force output (Gefen, Megido-Ravid, Itzchak & Arcan, 2001). Increases in postural sway, which often accompany fatigue, mark decreased stability. These increases in sway can be attributed to impairments of any of these systems (Lepers et al., 1997). As muscular fatigue occurs, postural sway increases, stability decreases, and the ability of the postural system to handle disturbances is inhibited (Yaggie and McGregor, 2002). The onset of muscular fatigue, thus, requires more exertion from the postural system in order to maintain correct posture (Corbeil et al., 2003). The existing literature has shown decrements in balance and postural control with progressive fatigue. But, these recent studies have investigated the effects of a relatively high level of fatigue by using eccentric contractions of greater than 50% of a particular maximal voluntary contraction (MVC) and with greater than 33% of maximum aerobic capacity (MAC) (Yaggie & McGregor, 2002; Vuillerme et al, 2002; Lepers et al, 1997). But, the workloads experienced in the occupational and industrial settings are usually different from that encountered during strenuous physical exercises. The intensity of these workloads are often relatively lower than the exercise induced workloads and often exposed over a prolonged duration rather than over a short and specific time period such as different exercise protocols. Workplace fatigue have been shown to be induced over a prolonged time period at a very low rate such as lesser than 15% MVC and lesser than 33% of MAC (Davidson et al, 2004).

There seems to be a lack of literature describing the influences an extended duration of workload on fatigue and, thus, postural control/balance. Cham and Redfern (Cham and Redfern, 2001), upon completion of their study on the effects of flooring on fatigue and standing comfort, recommended that similar studies be performed with at least a four hour duration to determine with optimum accuracy the effects of long duration walking and standing on muscle fatigue and postural control. Additionally, this study found that weight shifts at the center of pressure, indicative of decrements in balance, accompanied fatigue (Cham and Redfern, 2001). It was concluded that there was miniscule statistical significance in muscle fatigue until the third hour of standing, at the earliest (Cham and Redfern, 2001). Wade and Davis also found similar results with a prolonged exposure to an inclined surface and they recommended research on additional workload exposure time on balance (Wade & Davis, 2009).

The effects of footwear on balance:

The human locomotion is explained as the translatory progression of the body which is made possible by the coordinated rotatory movements of the body segments. The alternating movements of the lower extremities are responsible for carrying the head, arm and trunk complex along with them. Postural control can be defined as the ability to maintain stability of the body and its segments in response to the forces that disturb the body's equilibrium (Levangie & Norkin, 2006). And humans being bipeds and locomote with one feet on the ground in walking and with no feet on the ground in running and stand and assume a static posture with both feet on the ground (Winter, 1995), which is a challenging task for the postural control system. The assumption of a static posture is achieved by a relatively smaller base of support (BoS) formed by the area encompassed by both feet and the area between them. Since the foot is essentially the BoS for human balance, its stability is essential in the preservation of postural control. Due to the relatively smaller size of this BoS, even the smallest biomechanical alterations could have profound effects on the support (Cote et al, 2005).

Footwear serves as the interface between the human body and the supporting surface and along with the different features of shoe design such as the heel height, heel-collar height, sole hardness, heel and midsole geometry and slip resistance of the outer sole can significantly affect the balance outcome measures (Menant et al. 2008). About 45% of all falls have been attributed to inappropriate footwear (Gabell, Simons, Nayak, 1985; Menant, 2008). Footwear being the interface between the human body and the terrain play an important part in affecting balance and gait kinematics. Slight modifications in the mechanical characteristics of the shoes are compensated easily and automatically by the neuromuscular control of the individual. But, the modifications that are present on occupational footwear may be more pronounced for the neuromechanical adaptations to compensate for the loss of balance and in trying to maintain normal balance.

Different types of footwear affect gait and posture kinematics adversely. Improper alignment of the foot altered by different footwear leads to an increased metabolic cost, which in turn leads to a faster rate of development of muscular fatigue. Many literature and researches have analyzed gait and balance with different gait speeds, changing terrain, shoe types and in bare foot condition (Perry, Radtke & Goodwin, 2007, Menant, Perry, Steele, Menz, Munro & Lord, 2008, Divert, Mornieux, Baur, Mayer & Belli, 2005, Bohm & Hosl, 2010).

Based on the previous studies with CoP excursions, postural balance was shown to have a decrement with different footwear (Menant et al. 2008; Bohm & Hosl, 2010; Perry, Radtke & Goodwin, 2007; Hosoda et al, 1998). High shafts of the shoes / boots have an impact on the ankle range of motion which in turn leads to alteration in the power generation at the ankle joint for propulsion. The high boot shaft has the advantage of providing support and stability to the ankle (Bohm & Hosl, 2010). The influence of footwear midsole hardness was assessed by Perry et al. and they found that variations in the midsole material and even the presence of it impair the dynamic balance control system (Perry, Radtke & Goodwin, 2007). Body sway, maximal balance range, coordinated stability and choice-stepping reaction time were assessed by Menant et al and they reported that an elevated heel of 4.5 cm significantly impairs balance, whereas a hard shoe sole and a high heel collar may enhance balance in older people (Menant et al. 2008).

The central focus of this study is the effect of occupational footwear on balance with a prolonged period of exposure to walking and standing. The footwear of interest to the study are the Steel-toed work boot (WB). The WB met ANSI-Z41-1991 standards as per the OSHA regulations (Occupational Safety and Health, Laws and Regulations, 1970) for footwear in safety and protection, which are equipped with steel toes or metatarsal guards that provide toe protection from impact and compression injuries, oil resistant soles, and an elevated boot shaft height that extends above the ankle joint with distinct heels (Occupational Safety and Health Administration, U.S. Department of Labor). The Tactical boot (TB) has relatively lower shaft height to the WB, still extending above the ankle joint, with lower heel height and an athletic sole. The low-top slip resistant shoe has a lower shaft height than the WB and LB and does not extend over the ankle joint. It also has lower heel and a flat slip resistant sole. The LB and TB commonly used in a varied population for which there is not a prescribed OSHA regulation for the footwear to meet. These occupational footwear also have different masses which has been shown as an important mechanical characteristic that can affect balance.

The WB and TB are primarily designed for safety and protection whereas; the LB's design serves for comfort rather than safety and protection. The WB and TB with their elevated boot shafts may aid in balance by providing support around the ankle and thus decreasing the need for activation of additional ankle musculature and, thus, fatigue (Bohm and Hosl, 2010; Cikajlo & Matjacic, 2007). The characteristic elevated heel of these boots may also impair balance performance (Menant and Steele et al., 2008). Furthermore, the difference in the masses of these footwear may also influence balance and postural stability, as an increase in 100grams on the footwear has been shown to cause an increase in energy expenditure by 0.7% to 1.0% (Jones, Toner, Daniels & Knapik, 1984) and has been predicted that an heavier footwear may cause a decrement in balance performance (Chander et al, 2011). Hence the difference in the masses of these footwear may be an influencing factor in predicting balance performance.

Purpose of the study

Balance mechanisms during static and dynamic postures have been studied extensively and consequently, there have been several studies assessing balance in relation to different types of surfaces exposed such as types of flooring (Cham & Redfern, 2001), inclined surfaces (Wade & Davis, 2009), ballast (Anderes, Holt & Kubo, 2005) and in sand (Pinginton,0000). This study focuses on a hard firm floor as the exposure surface. Research has also focused on the type and intensity of workload that are placed on the human body that induce fatigue and its effects on balance (Gribble & Hertel, Yaggie & McGregor, Menant et al; Hosoda et al., 1998).

While extensive research exists on fatigue and balance, footwear and balance, terrain and balance, there has been less focus on extended durations of workload with different types of occupational footwear on balance on a hard firm surface. Therefore the purpose of the study is to analyze the effects of occupational footwear (WB, TB and LT) on balance, when exposed to a hard firm surface for extended durations of walking and standing.

Hypotheses

Balance Hypothesis:

Specific Aim 1:

To investigate the effects of extended durations of standing and walking on a hard firm surface 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_{01}: Individuals' balance will not be affected while exposed to extended durations of walking on a hard firm surface.

H_{A1}: Individuals' balance will be impaired while exposed to extended durations of walking on a hard firm surface.

The collaborative functioning of the vestibular, somatosensory, and visual systems is a direct determinant of the body's ability to maintain balance. According to earlier research on balance, decrements in any of these systems increase postural sway results and thus decrease stability. Furthermore, the research demonstrates that muscular fatigue of the postural muscles is accompanied by an increase in postural sway, as demonstrated by center of pressure weight shifts, indicating decreased stability. Additionally, the postural systems become less capable in compensating for disturbances with augmenting muscular fatigue, and eventually, more exertion is required from the postural system to sustain erect posture and maintain equilibrium. Other significant research asserts that progressive fatigue will occur with standing or walking duration. Therefore the alternative hypothesis is expected to be supported in this study, in that the exposure to extended durations of walking on a hard firm surface will increase the postural instability as measured in this study.

Footwear Hypothesis:

Specific Aim 2:

To investigate the effects of occupational footwear (WB, TB & LT) with extended durations of standing and walking on a hard firm surface 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_{02}: There will be no differences between different footwear conditions in individuals' balance while exposed to extended durations of walking on a hard firm surface.

H_{A2}: There will be significant differences among different footwear conditions in individuals' balance while exposed to extended durations of walking on a hard firm surface.

The feet, as the body's base of support and important sense organs in center of pressure adjustment, are indispensable in the balance and postural control system. Therefore, the stability of the foot is essential for balance maintenance. Furthermore, the type of footwear is an important determinant of postural control because it serves as the medium between the foot and support surface and affects somatosensory feedback mechanisms. According to the previous research, characteristics of footwear, such as midsole hardness, shoe elevation, heel elevation, boot shaft height, and boot shaft stiffness heavily impact postural control strategies and balance. Midsole hardness affects stability by center of mass fluctuations within the base of support and tactile sensory input transmission to sole receptors. Shoe and heel elevation also affect center of mass and pressure distribution. The boot shaft height and stiffness affect postural control by influencing ankle joint stability, range of motion, and fatigue. Because the different types of footwear utilized in this study (WB, TB & LT) have significantly different characteristics in these respects, the alternative hypothesis is expected to be supported in that there will be significant differences among the footwear conditions in individuals' balance during extended walking durations on a hard firm surface.

Operational Definitions

Posture:

Posture is essentially 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 may 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, Schwartz & Jessell, 2000).

Balance:

The ability to maintain the vertical projection of the center of mass within the base of support can be defined as Balance. Balance and postural stability are often used synonymously. Postural stability depends on the intentional action, the choice of movement strategy and the under lying neuromotor process (Levangie & Norkin, 2006).

The maintenance of the center of gravity within the limits of the base of support, which is determined by foot position (Kincl et al., 2002); the ability to maintain the center of mass over the base of support in order to sustain equilibrium in a gravitational field (Horak, 1987).

Fatigue:

Muscular fatigue can be defined as an inability of the muscle to maintain a reasonably expected force output (Gribble & Hertel, 2004). A cognitive perception of tiredness (Cham and Redfern, 2001). A decline in the capacity to generate force (Corbeil et al., 2003).

Electromyography (EMG):

Electromyography (EMG) is a clinical technique for evaluating and recording physiologic properties of the muscles at rest and while producing force. EMG is performed using an instrument called an electromyograph, to produce a record called an electromyogram. An electromyograph represents the spatial and temporal summation of all motor unit action potentials in the proximity of the recording electrode. It is indicative of the level of muscle activity via the motor unit recruitment and rate coding (Basmaijan, 1985).

Center of Mass (COM):

COM is defined as the point on a body that moves in the same way that a particle subject to the same external forces would move. It is also the point where the 3 mid-cardinal planes of the body meet. The center of mass is not necessarily located in the body (Rodgers & Cavanagh, 1984).

Center of Gravity (COG):

COG is defined as the point at which a single force of magnitude mg (the weight of the 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. In other words, the point at which the weight of the body or system can be considered to act (Rodgers & Cavanagh, 1984).

The center of gravity and the center of mass are coincident, although in strict physical terms, there is an infinitesimal difference between the two. The center of gravity of the human body is not fixed at an anatomical location. Its location varies according to the position of the body segments (Rodgers & Cavanagh, 1984).

Line of Gravity (LOG):

LOG is defined as the perpendicular line towards the ground from the center of gravity (COG) of that particular body (Levangie & Norkin, 2006).

Base of Support (BOS):

The human species' base of support (BOS), is defined by the area bounded posteriorly by the tips of the heels and anteriorly by a line joining the tips of the toes, and is considerably smaller than the quadruped BOS (Levangie & Norkin, 2006).

Center of Pressure (COP):

COP is defined as a quantity, available from a force platform describing the centroid of the pressure distribution. It can be thought of as (and is sometimes called) the point of application of the force (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 (NeuroCom International, Inc. Clackamas, Oregon).

Proprioceptive System:

The body system which promotes body position awareness and contributes to the maintenance of balance; includes input from the muscles, tendons, and joints; sensory receptors involved include those in muscle spindles, skeletal muscles, and Golgi tendon organs, which supply information on muscle length and tension, muscle force, and velocity (Sturnieks and Lord, 2008).

Somatosensory System:

The body system which includes the tactile and proprioceptive systems; includes input from Meissner's corpuscles, Pacinian corpuscles, Merkel's disks, and Ruffini endings, which all are touch inputs to the central nervous system (Hijmans et al., 2007).

Vestibular System:

The body system responsible for information including head position and motion relative to gravity, head posture, and body and eye movements; the structures of the vestibular system are in the inner ear (Sturnieks and Lord, 2008).

Visual System:

The body system which provides environmental information via the eyes as well as input about movements and position of the body; very important in posture in balance in that information from this system is used to regulate postural sway (Sturnieks and Lord, 2008)

CHAPTER II

REVIEW OF LITERATURE

The purpose of this investigation was to assess the effects of walking / standing for extended durations on a hard surface while donning different occupational footwear. This chapter will provide an insight to previous literature on postural control and balance in response to different types of footwear and to different types of workload (fatigue) placed on the postural control system. This chapter is divided into three major sections. The first section includes discussions about postural control and balance, the systems involved, strategies implemented and assessment parameters. This is followed by the final two sections, which comprises of a description and comparison of the relationship between balance and footwear; balance and workload (fatigue).

Postural Control and Balance

Posture can be defined as the orientation of the body or any of the body segments relative to the gravitational vector, whereas balance is a generic term that describes the dynamics of body posture to prevent falling (Winter, 1995). Balance and postural stability are often used synonymously. The ability to maintain the vertical projection of the center of mass within the base of support can be defined as Balance (Levangie & Norkin, 2006; Adlerton & Mortiz, 2003). Postural stability depends on the intentional action, the choice of movement strategy and the under lying neuromotor process (Levangie & Norkin, 2006).). In order to maintain balance in bilateral or unilateral stance, the center of gravity must be kept within the limits of the base of support, which is determined by foot position (Kincl et al., 2002). The limits of stability can be defined by the perimeter of the base of support in the anterio-posterior and medio-lateral directions. These represent the maximal excursions that the body can incur without falling (Yaggie & McGregor, 2002). The vestibular somatosensory and visual systems are responsible for identifying these limits of excursion and also continuously stimulate the muscles for corrections that are needed to reestablish balance (Yaggie & McGregor, 2002).

The maintenance of posture and balance are an important part of daily activities under both static and dynamic conditions. The afferent information from the somatosensory, vestibular and visual systems is processed in the brainstem and cerebellum after which the motor commands are initiated (Lepers et al. 1997). Of these three major sensory systems that maintain balance, the visual system is primarily responsible in planning our locomotion and in avoiding obstacles, while the vestibular system acts as a gyro, detecting linear and angular accelerations.

The somatosensory system is a proprioceptive system that senses the position and velocity of all body segments (Winter, 1995). 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).

The visual system contributes to balance through providing continuous information from the body's environment and supplies a feedback mechanism in body position and movement (Sturnieks and Lord, 2008). The visual system is integral during gait to foot placement and identification of surrounding hazards (Sturnieks and Lord, 2008). The maintenance of balance is dependent upon spatial perception as recognized by the visual system (Sturnieks and Lord, 2008). The vestibular system consists of structures in the inner ear which sense motion and position of the head in reference to gravity (Sturnieks and Lord, 2008). This aspect of the postural system also aids in posture and head, eye, and body movement coordination (Sturnieks and Lord, 2008). Recent studies have suggested that the vestibular system heavily influences fall risk (Sturnieks and Lord, 2008). The somatosensory system involves both the tactile and proprioceptive systems. Meissner's corpuscles, Pacinian corpuscles, Merkel's disks, and Ruffini endings all convey the sense of touch to the central nervous systems as a part of the tactile system (Hijmans et al., 2007). Furthermore, cutaneous mechanoreceptors in foot soles provide pressure distribution information to the central nervous systems. Information regarding joint angles and changes therein are provided by the proprioceptive system to the central nervous system, having been distinguished by Golgi tendon organs, muscles spindles, and joint afferents; however, it is currently unclear the role of feet and ankle proprioception in control of balance (Hijmans et al., 2007).

The CNS learns using the information from passive biomechanical elements, sensory systems and muscles. The CNS interprets and organizes inputs from various structures and systems and selects responses on the basis of past experience and the goal of response. Reactive (compensatory) responses occur as reactions to external forces that displace the body's COM. Proactive (anticipatory) responses occur in anticipation of internally generated destabilizing forces (Levangie & Norkin, 2006). Experiments are often performed involving the interference and conflict among the three systems to evaluate postural sway (Lepers et al., 1997).

The postural system must therefore 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 (Kandel, Schwartz & Jessell, 2000). The vertical projection of the center of mass (CoM); a point equivalent of the total body mass in the global reference system (GRS), on to 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 (Winter, 1995).

Upright maintenance of balance requires the individual to keep the center of gravity (CoG) within the base of support (BoS). The position of the center of gravity is changing constantly with the varying sensory input. The postural muscles are continuously working in order to keep the CoG within the BoS with minimal postural sway (Kincl, Bhattacharya, Succop & Clark, 2002).

Postural sway is usually described as a corrective mechanism in response to the external perturbations placed on the body. An inverted pendulum model is often used to describe postural sway, which bears a resemblance to a bilateral quiet stance, with ankle joint as the axis of rotation, along the sagittal plane. When the CoG vector is ahead or anterior to the CoP vector, the body will experience a clockwise angular velocity and angular acceleration, and in order to counteract this forward sway, a planter flexion moment or activation is performed. This will cause the CoG vector to be behind or posterior to the CoP vector, and causing a counter clockwise angular velocity and acceleration resulting in a backward sway of the body about the ankle joint. This is followed by the CNS sensing that the posterior shift in CoG needs corrective responses and decreases the plantar flexion moment or activation until the CoG lies ahead or anterior to the CoP. This sequence of events results in the postural sway of the body in the anterior-posterior (AP) direction about the ankle joint. The ankle plantar flexors and dorsi-flexors control the net ankle moment and thereby regulate the body's CoG (Winter, 1995). A similar inverted pendulum model can be suggested for the postural sway in the medio-lateral (ML) direction, with two ankle, knee and hip joints about the frontal plane. The AP control of postural sway is governed by collaborative 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 (Winter, 1995). 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 (Winter, 1995).

The ankle and the hip strategy are commonly used to control the body sway (Winter, 1995; Adlerton & Mortiz, 2003). The hip strategy is recruited if the ankle strategy is not sufficient to regain balance. In an ankle strategy, the body above the ankle joint is assured to move a one stiff segment which can be visualized as an inverted pendulum, whereas in the hip strategy a multi-chain model hinged at the hips have been noticed (Winter, 1995; Adlerton &

Mortiz, 2003). Regulating the relationship between the center of mass motion and the base of support is vital in maintaining balance. The rapidly generating muscle torques at the hips, ankle and other joints are responsible for decelerating the center of mass motion. And a greater degree of stabilization is achieved by rapidly changing the base of support, which can be done by initiating a step or by modifying a step, or by reaching to grasp or touch an object for support (Maki et al., 2008).

Assessment of postural balance is an indirect measure of the effect of both physiological and biomechanical stress on a worker and can be a very good outcome measure of the worker's overall safety status (Kincl, Bhattacharya, Succop & Clark, 2002). In an occupational setting the maintenance of upright balance is of utmost importance for the workers to perform the job tasks more efficiently and safely. In this setting, postural instability can be hazardous due to the increased risk of falls/slips and other accidents (Kincl, Bhattacharya, Succop & Clark, 2002). Increased postural sway which is regarded as a decreased stability is due to the reduction in peripheral sensibility in the visual, vestibular or proprioceptive systems (Lepers et al. 1997). It can also be due to a defect or a slowed response of the central interrogative mechanisms responsible for configuring the postural control systems. An increase in postural sway can be an indicator for impairment of postural control as a result of functional instability (Yaggie & McGregor, 2002) and an increased medio-lateral sway is strongly associated with increased lateral instability and thereby an increased incidence in fall rates (Hijmans, Geertzen, Dijkstra & Postema, 2007).

The fundamental prerequisite for a fall includes; an initial loss of balance induced by a perturbation such as a slip, trip, misstep or a collision and a failure of the balance recovery mechanisms to counteract the destabilization (Maki et al., 2008). This is more pronounced in an industrial setting due to hazardous environment such as slippery surfaces, trip hazardous and added with the need to perform distracting or destabilizing tasks while standing or moving (Maki et al., 2008).

There have been various procedures employed to quantify postural control such as using force plates and accelerometer (Adlerton & Mortiz, 2003) and using time-to-boundary measures using CoP patterns (Hertel, Olmsted-Kramer & Challis, 2006). Questions regarding the effects of sampling duration or sampling frequency were assumed by Winter (Carpenter, Frank, Winter & Peyser, 2001) where the impact of sample duration on the magnitude and reliability of the Center of Pressure summary measures in both the frequency and the time domain were observed over an interval of 120 seconds. He suggested that sample duration of at least 60seconds should be used to optimize the stability and reliability of the root mean square (RMS) summary measures of CoP during quiet stance (Carpenter, Frank, Winter & Peyser, 2001). By considering the human body as a relatively rigid body the center of gravity’s motion corresponds to the displacements of the whole body which serves as an evaluation tool for postural stability on net performance. Force plates have been used to assess the projection of the center of mass in quite standing (Caron, 2002). The changeable positions of the CoP are registered and calculation of CoP velocity and amplitude are made possible. Decreased stability or poor balance can be defined with an increased measure of the CoP velocity and amplitude. Choice of strategy and movements of a reference point on the body can be studied by an accelerometer (Adlerton & Mortiz, 2003). The most common measures used for the assessment of postural stability are the CoP excursions. Center of pressure excursion velocity and area determine the amount of decline in the postural control. A novel approach for this assessment was the time to boundary (TTB) measures of the center of pressure excursions (Hertel, Olmsted-Kramer & Challis, 2006). Hertel et al. described

TTB as an estimate of the time it takes for the center of pressure to reach the boundary of the base of support, if the center of pressure were to continue on its trajectory at its instantaneous velocity. This gives an insight to the spatio-temporal characteristics of postural control. A lower time to boundary measure indicates a propensity to postural control instability because of the reduced time available to execute a postural correction which was also comparable to the CoP measures (Hertel, Olmsted-Kramer & Challis, 2006).

The contribution of each sensory input toward equilibrium can be ascertained by measuring the equilibrium adjustments of a standing subject, where input from the visual; support surface and environment are recorded by presenting conflicting visual, proprioceptive and vestibular stimuli. Thus, the experiment of dynamic posturography involves creating various conflicting sensory conditions by rotating the surface platform and/or the visual surrounding in proportion to the subjects postural sway. This has been used extensively in assessing human balance and posture, especially in clinical practice to differentiate disturbances if vestibular, visual and proprioceptive functions, including central coordination (Lepers et al. 1997).

The sensory organization test (SOT) on the Neurocom Equitest evaluates the integrity of the three sensory modalities by selectively disrupting somatosensory and/or visual information regarding the CoG orientation in relation to vertical and then measuring the individual's ability to maintain balance (Guskiewicz & Perrin, 1996). The sway referencing capabilities of the support platform and the visual surround involve tilting in the AP direction and responsible for maintaining the orientation of the support platform and the visual surround constant in relation to the CoG sway angle (Guskiewicz & Perrin, 1996).

Balance and Foot wear

The human foot is the first point of contact between the body and the environment or terrain which is vital in relaying the somatosenroy information to the CNS both during static and dynamic balance tasks. Furthermore, footwear serves as the interface between the human body and the supporting surface and can significantly affect the balance outcome measures (Menant et al. 2008). Efficient transformation of the mechanical power output produced by the musculoskeletal system through the footwear is responsible for a good performance in gait. Hence, the design and type of the footwear becomes important in gait and posture (Bohm & Hosl, 2010). Forty-five percent of falls have been attributed with inappropriate footwear. Walking bare foot has also been related to an elevated risk of falls. The different features of the shoe design, such as the heel height, heel-collar height, sole hardness, heel and midsole geometry and slip resistance of the outer sole have been known to have on the influence on balance maintenance (Menant et al. 2008). Certain commonly worn footwear, such as slippers were found to be hazardous as they slowed down reactions to perturbations and also had adverse effects on posture reactions (Hosoda et al., 1997) with even, barefoot walking shown to lead to an increased risk of falling (Menant et al. 2008).

Energy expenditure has been shown to differ based on the viscoelastic properties and weight of the shoe. High shafts of the shoes/boots have an impact on the ankle range of motion which in turn leads to alteration in the power generation at the ankle join for propulsion (Bohm & Hosl, 2010). The high boot shaft has the advantage of providing support and stability to the ankle. The biomechanical function of the boot shaft is to restrict inversion and thereby protect the ankle from sprains such as the lateral collateral ligament sprain of the ankle, which is one of the most common sprains encountered. A considerable boot shaft thickness is required to perform this function effectively (Bohm & Hosl, 2010). The influence of footwear midsole hardness was assessed by Perry et al. the measurement of bare foot condition was treated as a control. They tested 12 healthy young female adults aged 20-23 years. The primary outcome measures were the Maximum and Minimum range of the transverse plane projection of the center of mass location relative to the lateral base of support (CoM-BoS), Maximum and Minimum of the center of mass and center of pressure (CoM-CoP) difference in the anterior-posterior direction and the average vertical force loading rates. They found that variations in the midsole material and even the presence of it impair the dynamic balance control system (Perry, Radtke & Goodwin, 2007). Balance assessments were done by Menant et al. on 29 community dwelling volunteers of 70 years of age and older by measuring body sway, maximal balance range, coordinated stability and choice-stepping reaction time. The findings from the study reported and confirmed that an elevated heel of 4.5 cm significantly impairs balance, whereas a hard shoe sole and a high heel collar may enhance balance in older people (Menant et al. 2008). Cowboy boots with an elevated heel was found to have decreased balance performance compared to tennis shoes in a population of 27 healthy women of 18 to 40 years of age (Brechet et al. 1995). Further research on specific boot characteristics that might have contributed to a decreased balance performance was warranted.

Dynamic balance was assessed among 43 healthy university students with different footwear such as slippers with and without clog thongs, leather soled sandals and Japanese socks. Decreased postural response latencies to horizontal movement of platform and decreased standing strength when the platform moved horizontally were seen in footwear with clog thongs when compared with the one without clog thongs (Hosoda et al., 1998). Studies on balance enhancing footwear insoles and their effects on stepping reactions (Maki et al., 2008) and gait patterns (Nurse et al. 2005) are well documented. There have been studies that looked at the effects of shoe characteristics on the dynamic stability among young and old population on even and uneven surfaces. A conservative walking pattern was observed with elevated heel shoes and a decrease in the medial-lateral balance in soft sole shoes among both the populations. Walking stability was not seen to improve in either group with increased sole hardness, a tread sole and a raised collar height shoe, when walking on even as well as uneven surfaces (Menant et al. 2008).

Balance and gait parameter in older women were assessed in barefoot, and while wearing walking shoes and dress shoes. The elder women had a better balance barefoot or with walking shoes in the functional reaching tasks in comparison to the dress shoes. The dress shoes also had the slowest gait speed followed by barefoot and walking shoes when assessed with the timed up-go test and 10 meter walk test (Arnadottir & Mercer, 2000).

Boots with increased shaft thickness such as the military boots have been shown to decrease the peak dorsiflexion of the ankle up to 4 degrees and a reduced peak power production at the ankle joint of about thirty-three percent compared to that of a soft boot shaft. Hosl and Bohm (Bohm & Hosl, 2010) found that the ankle joint had reduced power which was compensated by increased hip moments, which were needed to change the gait pattern from a push off mode to a pull off mode. They found that, the eccentric energy of the knee joint increased and the eccentric energy in the ankle joint decreased when wearing a harder boot. They couldn't identify if the additional effect on the knee joint is an effect of the uneven surface, amplifying possible compensation mechanism, or due to the extended measures of the actual study. Output measures included EMG and joint energy from the limb tested. They warranted further research comparing the uneven surface with the level ground surface (Bohm & Hosl, 2010). The differences in boot shaft and vamp stiffness was also assessed by Cikajlo and Matjacic, in which the softer boot shaft enabled a greater range of motion and a greater power generation in the ankle joint during push off (Cikajlo & Matjacic, 2007) which supports earlier studies.

Kinematics analysis of military boots in comparison to flip flops and barefoot showed an increased step length and stride length with the military boot with a significant reduction in cadence. This was supported by an increased time period in swing phase and single limb support; with a decreased total support time and initial double support time in the military boots (Majumdar et al. 2006). An increase in the distal mass with a pendulum lengthening effect on the leg and hence an increased inertia during swing phase has been postulated as potential mechanisms for an increase in stride length and cadence (Majumdar et al. 2006). Although stance phase duration decreased in the footwear conditions in comparison to barefoot, the difference in heel height did not affect the stance phase duration, with the differences primarily existing in the pressure distribution on the sole of the foot (Eisenhardt et al. 1996).

Balance and Fatigue

Low level of muscular forces are necessary for stabilizing the center of mass over the base of support and in erect bipedal stance the base of support is a very small area covering the two foots and the area between them (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003). The base of support is even more considerably reduced in unilateral stance. Muscular fatigue may impair the proprioceptive and kinesthetic properties of joints by increasing the threshold of muscle spindle discharge, disrupting afferent feedback and ultimately altering conscious joint awareness (Gribble & Hertel, 2004).

Fatigue can be considered as internal perturbation which tries to displace the body posture away from equilibrium by destabilizing the body's COM (Nardone, Tarantola, Giordano & Schieppati, 1997). The CNS ability to anticipate minimal body destabilization occurs as a function of the proprioceptive, visual and vestibular systems. When these are held defective, body sway increases and muscle activity increases to maintain balance. Further, sufficient ankle power is necessary for the forward motion during gait to maintain a normal walking velocity. And a reduced power generation at the ankle joint can impair the stability during gait and during static stance. This may influence compensatory changes at the knee and at the hip joints (Bohm & Hosl, 2010).

The torques produced by external perturbations leading to store destabilization are large enough that the mechanical characteristics of the muscle are not sufficient to compensate and as a result, balance is actively controlled by the CNS where postural muscles are recruited whenever needed (Nardone, Tarantola, Giordano & Schieppati, 1997). After a fatiguing exercise of the postural muscles, the gastrocnemius and soleus, if a change from an ankle strategy to a hip strategy is observed, it can be attributed to the decreased frequency of the COP as there is an increased mass involved in controlling balance and also because there is an increase in trunk acceleration due to more active role of the hip and trunk. The fatigue was thought to put more stress on the postural control systems (Adlerton & Mortiz, 2003). There have been many observations that equilibrium is impaired after prolonged exhausting physical exercise (Pline, Madigan & Nussbaum, 2006, Caron 2004, Yaggie & McGregor, 2002, Gribble & Hertel, 2004). This also holds true for prolonged or extended durations of walking or standing. The muscle spindles, tendon organs, joint receptors and cutaneous afferents on the sole of the feet are shown to be activated with each stride. The vestibular system was shown to be sensitive during head accelerations and the eyes being constantly stimulated by the moving visual fields (Lepers et al. 1997).

The ability to evaluate joint position, movement direction and speed are crucial factors in maintaining balance. Minor perturbations are often taken care of by the ankle response which co relates with the proposed idea of humans behaving like inverted pendulums. However, perturbations such as fatigue and vibrations are known to affect and impair postural control (Vuillerme, Danion, Forestier & Nougier, 2002). With muscle fatigue, the deterioration of postural control can be related to the deterioration of the inability of the muscles to produce and sustain a required output and also due to the reduced activity of the proprioceptive system (Vuillerme, Danion, Forestier & Nougier, 2002). Localized muscle fatigue may affect the control of balance and posture (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003). Multiple sensory systems and motor components of the nervous system are involved during the control of posture and balance. When the sensory or motor components are altered or defective, body sway generally increases and muscle activity increases concurrently, in order to maintain postural equilibrium (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003).

Various fatigue models have been suggested with regard to different levels of the nervous system. At the peripheral level, a failure of the muscle to respond to neural signal or a failure of the muscle to respond to neural excitation can be attributed to muscular fatigue. And at the central level, fatigue is known to induce a failure of excitation of the motor neurons caused by changes in the nervous system. The changes in the motor neuron firing has been attributed to the intrinsic properties of the motor neurons, recurrent inhibition due to the Renshaw cells and changes in reflex inhibition or due to changes in the descending drive in the motor neuron pool (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003). Perturbations caused by the muscle fatigue in regard to joint position sense have been related to decreased motor neuron output, or desensitization of the type 3 and 4 muscle afferents (Yaggie & McGregor, 2002).

Muscular fatigue represents an unavoidable occurrence for physical, work related, and daily activities that the CNS has to take into account (Vuillerme et al., 2002). Fatigue has been suggested to negatively affect the propriocetive system through either deficiency in the activation of the muscular mechanoreceptors or a decrease in the muscular function (Vuillerme et al., 2002; Corbeil et al., 2003). Increases in postural sway accompanied by localized muscle fatigue indicate an impairment of postural control, which are usually associated with an increase in fall rates. Falls are more prone to happen in the occupational environments with increase in fatigue (Pline, Madigan & Nussbaum, 2006). In these circumstances, localized muscle fatigue in induced at a relatively low threshold over a period of several hours. Fatigue time and the amount of fatigue were considered as important factors influencing the postural sway and thereby influencing the postural stability (Pline, Madigan & Nussbaum, 2006). The effect of prolonged exercise has been well studied in terms of musculoskeletal fatigue but not from the neuro-sensory system perspective and regarding the effects of fatigue on the maintenance of equilibrium and balance (Lepers et al. 1997). The maintenance of an upright posture is necessary for the workers in their occupational environment. Maintaining postural balance for workers is important in order to perform tasks or jobs safely. Higher incidence of falls, decreased balance measures and the ability to perform tasks within it accompanied by fatigue is a hazardous condition for workers (Kincl, Bhattacharya, Succop & Clark, 2002).

The effect of fatigue on decreasing postural stability has been reported quite often (Caron, 2002, 2004, Yaggie & McGregor, 2002). But there have been contradictory results whether or not visual-sensory input may compensate for the destabilizing effects of fatigue (Caron, 2004, Vullerme, Burdet, Isableu & Demetz, 2006). Vullerme tested 12 healthy subjects using a force platform before and after exercises on the calf muscle with three different visual conditions of vision, no vision and with vision of a block cross placed at a distance of 4 m ahead. He concluded that there was a decreased postural control following gastrocnemius and soleus fatigue during quiet biped standing in the absence of vision and the ability to use visual information to compensate for this destabilizing effect was dependent on the eye-visual target distance (Vullerme, Burdet, Isableu & Demetz, 2006). This also proved good for the “Sensory-Motor adaptation process” where any deficit in one sensory modality is often compensated for by the enhancement of the sensory weights of all other intact sensory modalities based upon the relevance of other sensory cues in a given environmental context.

Nordone, (Nardone, Tarantola, Giordano & Schieppati, 1997) assessed the effects of fatigue involved by treadmill walking and cycle ergometer pedaling on balance and posture, measured by body sway area and sway path among 13 healthy young subjects. They found that fatigue had a significant effect on body sway variables. However the cycle ergometer was found to have a swollen effect on the sway variables when compared to the treadmill. They concluded that strenuous exercise does indeed affect body balance during maintainance of quiet upright posture but the consequent increase in sway was short lasting and of moderate extent and therefore were not liable to seriously threatening body equilibrium (Nardone, Tarantola, Giordano & Schieppati, 1997). Fatiguing protocols usually followed experimental setups involve cycling and running isometric fatiguing protocol of the lower limbs or with MVC of one of the postural muscles (Caron, 2002). But inducing fatigue has never been done in a way that is as close as possible to an occupational setting. Postural stability was assessed by the COG motion, which was computed from the motion of the center of pressure, evaluating the postural control. EMG analysis of the tibialis anterior and soleus muscle were also recorded from the 10 healthy male subjects. They concluded that there was modified postural control due to fatigue but did not modify much postural stability (Caron, 2002).

Vuillerme, (Vuillerme, Danion, Forestier & Nougier, 2002) tested the postural sway under muscle vibration and muscle fatigue. This experiment was in accordance with the hypothesis that muscle fatigue caused an increase in postural sway fatigue but also found that muscle vibration did not induce a further increase in postural sway. They concluded by saying that fatigued muscles may be less sensitive to muscle vibration and to some extent the CNS may decrease the valiance on proprioceptive information from the ankles and may use other sensory inputs providing more reliable information for regulating postural sway (Vuillerme, Danion, Forestier & Nougier, 2002).

Yaggie, (Yaggie & McGregor, 2002) used isokinetic contractions of plantar flexors and dorsi flexors to induce fatigue on twenty healthy men. The Anterior-Posterior and Medial-Lateral sway displacement were analysed immediately before and after posterior fatigue with 10 minute time intervals. Isokinetic fatigue of ankle plantarflexors and dorsiflexors were found to significantly affect sway parameters. It was shown that small perturbations during quiet stance were altered by ankle strategy by means of the stretch reflex. But when fatigue was introduced this ability to tolerate small disturbances in balance becomes more difficult. This was related to an impaired ability to reproduce lower extremity joint angles after a fatiguing protocol. These effects were attributed to the decline in proprioceptive joint function due to fatigue (Yaggie & McGregor, 2002). The effect of fatigue on postural control was shown by studying women between 20-34 years of age free from any injuries. Fatigue was induced with repeated heel raises until exertion and also reported by the values of the Borg's scale Rate of Perceived Exertion. It was shown that fatiguing exercise was responsible for a short lasting effect from the force plate and accelerometer measures between the fatigued and non-fatigued limbs. Trunk acceleration in the anterior-posterior and medial-lateral directions along with the center of pressure amplitude increased as a response to fatigue. Fatigue was responsible for a change in the postural control pattern relying on compensatory corrections from in and around the hip. Fatigue was shown to have altered the postural control mechanisms and had deleterious effects on maintenance of an upright posture (Adlerton & Mortiz, 2003). Lepers et al showed altered balance ability among 9 well trained athletes after inducing fatigue with prolonged running (Lepers et al. 1997).

Twelve physically active men aged 20-22 years, were tested by Pline et al, (Pline, Madigan & Nussbaum, 2006) with each performing multiple sets of back extensions with systematic adjustments of the number of repetitions in each set. This was done in order to achieve a specific amount of level of fatigue over a specific fatigue time. COP based measures of postural sway with mean velocity, peak velocity and sway area were used as testing variables.

Pline et al found that immediately after the fatigue protocol, postural sway was affected. They reported larger increases in sway velocity and sway area when fatigue was induced over longer durations and larger increases in sway velocity at higher fatigue levels. In the experiment conducted by Corbeil et al, (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003) muscular fatigue was induced with repeated plantar flexion of both legs among 11 healthy male subjects and postural stability was assessed in conditions with and without vision over 60 second period. An increased postural sway was observed in both conditions of eyes closed and open with no significant difference in the range of oscillation and the variability of the postural oscillations around the mean positions of the center of pressure. They concluded that, compared to the no-fatigue conditions, fatigue placed higher demands on the control of posture by increasing the frequency of actions that are needed to regulate the upright stance (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003).

All these recent studies have investigated the effects of relatively high rates of fatigue on balance [>50% Maximum Voluntary Contraction (MVC) and/or over >33% Maximum Aerobic Capacity (MAC)] over a short period of time (<10 min) period of time (Vuillerme et al., 2002; Yaggie and McGregor, 2002). These fatigue effects were studied using different types of strenuous physical exercises, such as cycling, running, or isometric or isokinetic fatiguing protocols of the lower extremities. However, in the workplace, fatigue may be induced at a lower rate of exposure (<15% MVC and/or <33% MAC) over a long period of time (1-8h) (Davidson et al., 2004). Although maintenance of upright stance does not require great physical effort, it is nonetheless a well coordinated task that can be impaired by minor losses in postural muscle force. Some tasks require intense physical effort to maintain postural control during work related activities, and fatigue can accrue rapidly in a short period of time. Postural fatigue may amass slowly in even light effort tasks due to static loading of the postural control measures, resulting in a decreased muscular force capacity. This reduction in the muscular force has been suggested to advance to a decrease in working capability and resulting in an internal perturbation to the motor control system and thereby impairment in motor coordination and quite possibly, in the postural control system (Nardone et al., 1998).

Previous literature exists on the assessment of balance and fatigue over an extended duration of a workload. The perturbations of equilibrium following a prolonged exercise were investigated by Lepers et al. using the SOT. The prolonged exercise protocol used was a 25 km run with an average time of 1 hour-45 minutes and a cycle ergometer of equivalent time period to the run. Postural stability reported with equilibrium scores form the SOT was found to decrease following the prolonged exercise protocol for both types of exercise with conflicting sensory input (Lepers et al. 1997).

In a more ergonomic setting, Wade, Weimar & Davis, assessed the influence of walking on a pitched roof setting for prolonged durations. Postural stability was found to decrease with increased sway velocities following exposure to the inclined surface (Wade, Weimer & Davis, 2003). Following this, Wade and Davis found increased sway RMS and sway velocities following an extended duration (2 hours) of exposure to an inclined surface. The authors suggested that this increased sway parameters was due to fatigue caused by the prolonged workload of walking on an inclined surface, although this was not substantiated with corresponding physiological measure of fatigue (Wade & Davis. 2009). The assessment of balance and fatigue over a 4 hour standing exposure to seven different flooring conditions was done by Cham and Redfern, in which the CoP shifts between each lower extremity was found to increase over the 4 hour testing duration. But significant differences were found only after the 3rd

hour of the testing time. Physiological and subjective measure of fatigue was done by EMG and a CR10 Borg scale respectively. Although the subjective measures of fatigue and discomfort increased over time, EMG findings were not sensitive enough to allow detection of muscular fatigue (Cham & Redfern, 2001). The authors also suggested that a minimum of 4 hours of exposure to standing and walking was needed to assess the influence of low intensity workload on postural control and balance.

CHAPTER III

METHODS

The purpose of this study was to examine the effects of wearing three types of commonly used footwear in the occupational industry while standing and walking on a hard firm surface for an extended duration of time (4 hours). Specifically an analysis was conducted to assess the effect on balance in relation to the footwear and with extended durations of exposure.

Participants

Fourteen healthy adult males were recruited based on an anthropometric blocked assignment for participation in this study. Written informed consent was obtained as per the regulation of the Institutional Review Board. Exclusionary criteria included orthopedic, musculoskeletal, cardiovascular, pulmonary and neurological abnormalities including vestibular diseases and any other difficulties in standing and walking that would hinder normal balance and/or gait and the successful completion of the testing session. Participant demographics are listed in Table 1.

Table 1

Subject Demographics	Mean ± SD
Age (years)	23.6 ± 1.2
Mass (kg)	89.2 ± 14.6
Height (cm)	181 ± 5.3

Instrumentation

Standing balance was assessed by the NeuroCom Equitest Balance Master - Posture Platform (NeuroCom International, Inc. Clackamas, Oregon) at the Applied Biomechanics Laboratory (ABL) in the Department of Health, Exercise Science and Recreation at the University of Mississippi. The system utilizes a dynamic 18” x 18” dual force plate with rotational and transitional capabilities and a visual seen with transitional capabilities encompassing sway referencing capabilities. The Sensory Organization Test (SOT) was used as the assessment tool for measuring balance. The experimental conditions of the SOT utilize the sway referencing capabilities of the platform and the visual seen to produce six conditions: standing with (1) eyes open (EO) and (2) eyes closed with the platform and visual surround stable (EC), (3) standing with the platform stable, eyes open with the visual surround sway referenced (EOSRV), (4) standing on the platform sway referenced w/ eyes open (EOSRP), (5) standing on the platform sway referenced w/ eyes closed (ECSRP), and (6) standing on the platform, eyes open, with the platform and visual surround sway referenced (EOSRVP). Foot forces were recorded to estimate center of pressure for sway analysis. If the need arises, a harness system was provided to prevent injury from falling during testing.

Experimental Procedure

Experimental Conditions

Participants were tested in three different conditions wearing three different types of footwear such as the Steel-Toed Work Boots (WB), Tactical Boots (TB) and Low Top-Flat Sole Slip Resistant Boot (LT). The WB met ANSI-Z41-1991 standards as per the OSHA regulations for footwear in safety and protection, which are equipped with steel toes or metatarsal guards that provide toe protection from impact and compression injuries, oil resistant soles, and an elevated boot shaft height that extends above the ankle joint with distinct heels (Occupational Safety and Health Administration, U.S. Department of Labor). The LT and TB commonly used in a varied population for which there isn't a prescribed ANSI standard for the footwear to meet. The average size of the footwear used in the study was a foot size of 11, for which the footwear characteristics are listed in the Table 2.

Table 2

Footwear Characteristics

Shoe	LB	TB	WB
Mass (kg)	0.4	0.5	0.9
Boot Shaft Height (cm)	9.5	16.5	18.5
Heel Sole Width (cm)	8.5	8.8	9.6
Forefoot Sole Width (cm)	10.5	11.0	12.0
Heel Height (cm)	2.1	3.5	3.8

Experimental Testing

The testing procedure was done in the premises of The Applied Biomechanics Laboratory (ABL) at the University of Mississippi. The testing procedure for each subject followed a repeated measures study design with duration of exposure to the hard firm surface as a nine level independent variable: Pre, 30, 60, 90. 120, 150, 180, 210 and 240 min of exposure time intervals while wearing the three types of footwear. The participants read and signed the informed consent after which they filled out a preliminary medical questionnaire. The testing procedure consisted of assessing the participants on the six conditions of the SOT (EO, EC, EOSRV, EOSRP, ECSRP and EOSRVP) on the NeuroCom Equitest. The first visit was treated as a familiarization period, where subjects were exposed to the SOT. The next visit was treated as the first condition, where each participant was randomly assigned one of the three footwear tested. The subjects were assessed on the NeuroCom, prior to the beginning of the walking session as a pre-test measure (Pre) and then again every 30 minutes for the entire 4 hours until the 240^{th} minute. The testing time intervals for all the three different types of footwear were as follows: Pre, 30 min, 60 min, 90 min, 120 min, 150 min, 180 min, 210 min, and 240 min. The exact same protocol was repeated for the following two subsequent visits for each participant with the other two remaining footwear which were presented in a randomized fashion. Participants were instructed to walk with a self selected pace and a self selected path on the hard firm surface until every 30 min intervals to complete the balance testing session. Participants were given at least 72 hours of rest between the testing conditions and were asked to refrain from exercising their lower extremities at least 48 hours before a testing session.

Data processing

The values of the dependent sway variables were derived from the Center of Pressure (CoP) movement, which were calculated from the raw data from the NeuroCom Equitest Balance Master. The average sway velocity (VEL) and the root-mean-square (RMS) of the CoP were used to characterize the postural sway in the anterior-posterior and the medio-lateral directions during the 60-second testing period. Velocity is determined by calculating distance over time and sway velocity in particular is a measure of the peak to peak change of the CoP per unit time. The RMS which is used as a rectifying measure estimates the amplitude of sway and the overall amount of movement of the CoP during the entire testing time on the Neurocom Equitest. Hence, the outcome variables were labeled as VEL and RMS in the anterior-posterior direction (APVEL & APRMS) and the VEL and RMS in the medio-lateral direction (MLVEL & MLRMS). Postural sway is induced as result of the constant adjustments and activation of the postural muscles in an attempt to keep the CoP within the center of the base of support (BoS). VEL determines the rate of this compensation and RMS gives the amount of compensation needed to maintain the CoP within the center of the BoS and thereby effectively maintain balance. Higher values of VEL and RMS indicate decreased postural stability and balance, as they imply larger angular changes in the location of the CoP. These two measures were used since they show different characteristics of postural sway. The Sway RMS and VEL were calculated using the following equations respectively.

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

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

Statistical analysis

A repeated measures analyses of variance (RMANOVA) was performed to determine if differences existed between boots type (WB, TB & LT) on postural stability measures over time. Postural stability dependent variables were evaluated using a 3 x 9 (Boot [WB v. TB v. LT]) x (Extended duration of walking intervals [Pre, 30, 60, 90, 120, 150, 180, 210 & 240] RMANOVA and independently for the six NeuroCom Equitest System testing conditions (EO, EC, EOSRV, EOSRP, ECSRP and EOSRVP) to identify any existing differences within the exposure time as well as the shoe types. If significance was found a pairwise comparison with a Bonferroni correction was done to determine among which time point the significance existed. A Greenhouse- Geisser correction was used to determine significance, if the Mauchly's test of sphericity was violated. For all analyses, significance was set at an alpha level of $p \leq .05$ and all statistical analyses were run using the SPSS 17 statistical software package.

CHAPTER IV

RESULTS

Participant information:

Fourteen healthy male adults completed the study successfully. Initially, sixteen individuals were recruited. One of the participants failed to complete the study and another participant was excluded from the data analysis due to erroneous data and frequent falls during balance testing.

Data analysis:

The sway parameters (VEL and RMS) calculated from the raw data were used for data analysis. An average value for the VEL and RMS was calculated from the three trials performed for each of the six testing conditions on the Neurocom Equitest. Falls that occurred during any of the particular trials while performing the SOT were excluded from the data analysis. The data was further winsorized at 5^{th} and 95^{th} percentile. Winsorization is the transformation of statistics by limiting extreme values in the statistical data to reduce the effect of possibly spurious outliers rather than excluding the data altogether.

Anterior-Posterior Sway RMS (APRMS)

A repeated measures ANOVA with a Greenhouse-Geisser correction determined that there were no statistically significant differences between the time points for all the six conditions of the SOT with Time as the main effect and no significant difference with the Time-Shoe Interaction. However, significant difference existed in the Eyes Closed Condition ($P = 0.007$) ($F(2, 22) = 6.238$, $P < 0.05$) with shoe type as the main effect. Post hoc tests using the Bonferroni correction revealed that the LB had a significantly greater sway RMS in the anterior-posterior direction when compared to both the TB and WB. There was no statistically significant difference among the TB and WB for the above mentioned condition.

Figures 1-6: Averaged Sway RMS measures in the Anterior-Posterior directions for each of the six postural stability testing conditions. # indicates a significant difference over time intervals and * indicates a significant difference between the boot types and the bars represent the standard errors.

Figure 1:

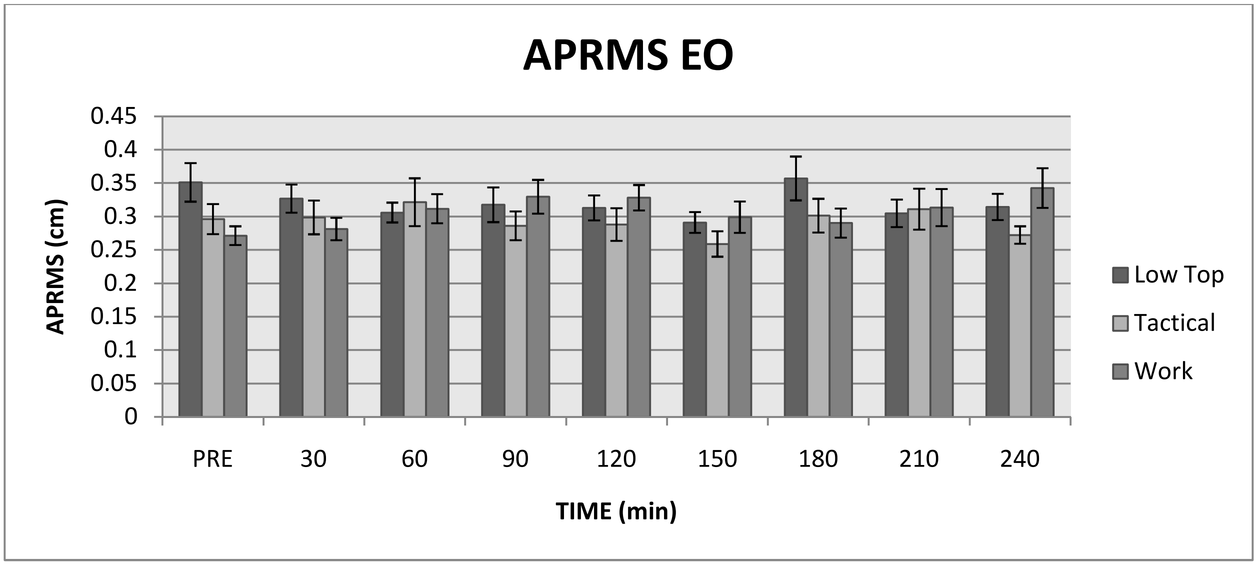
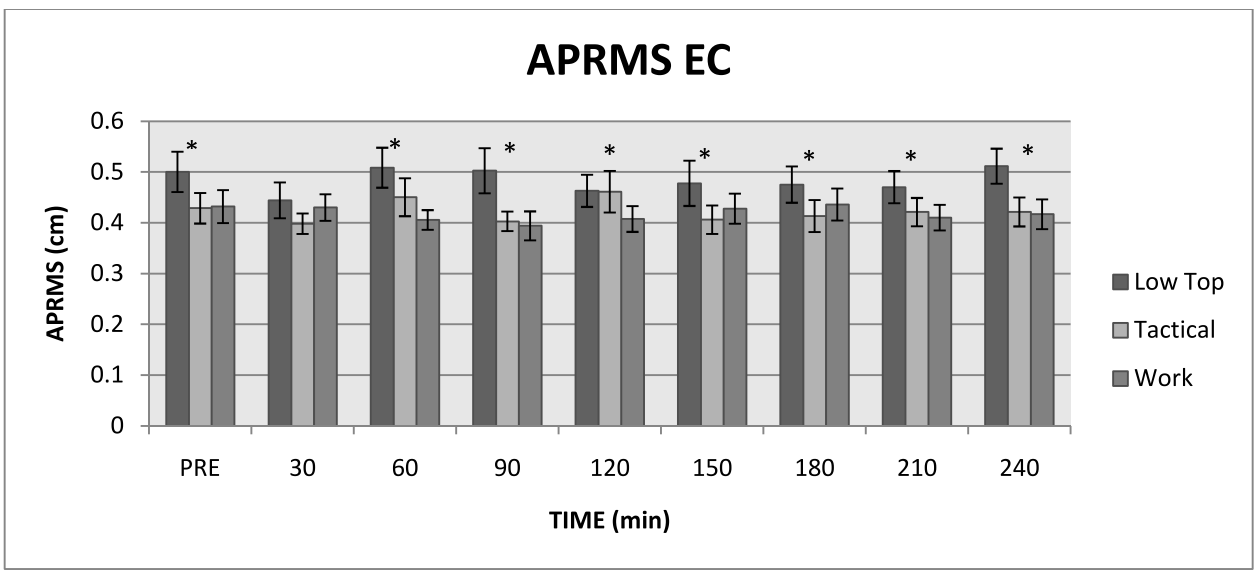


Figure 2:



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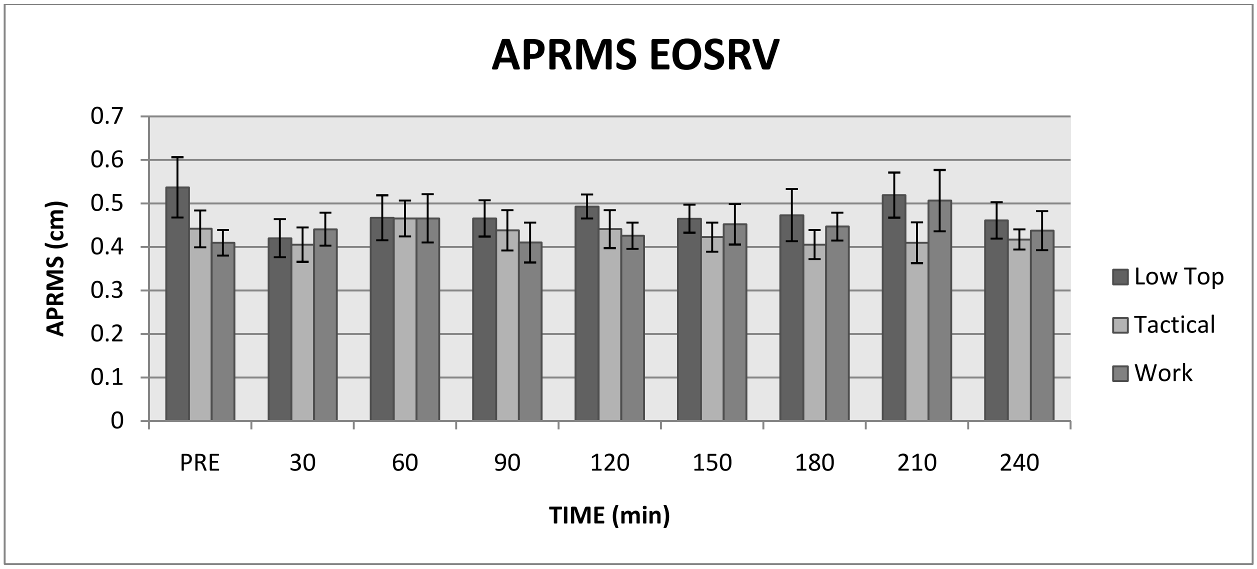

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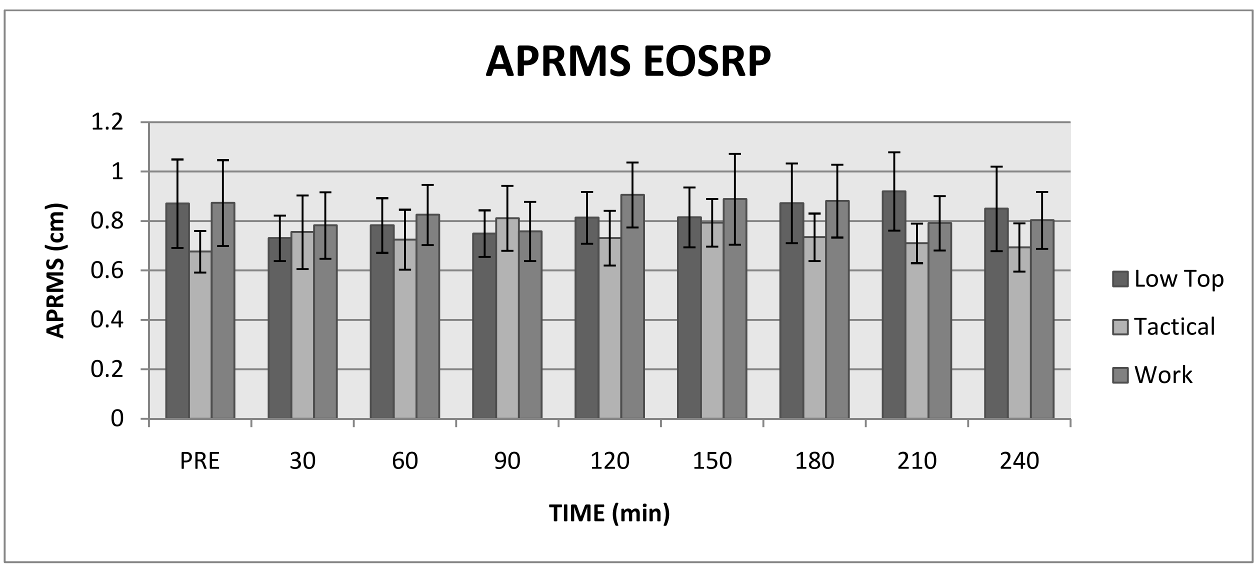

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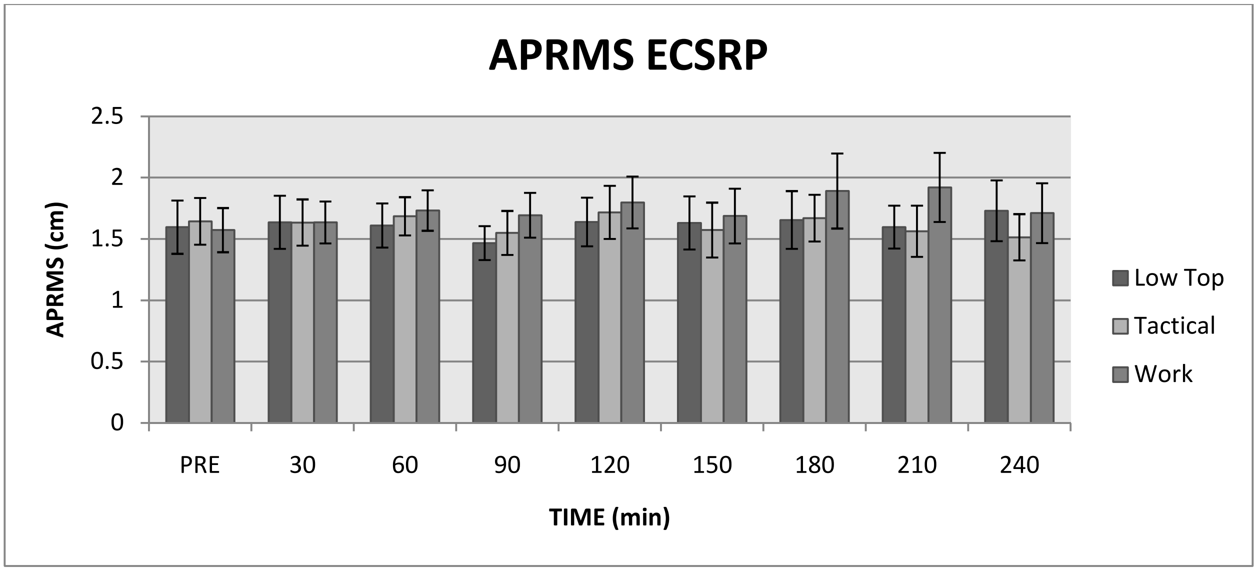
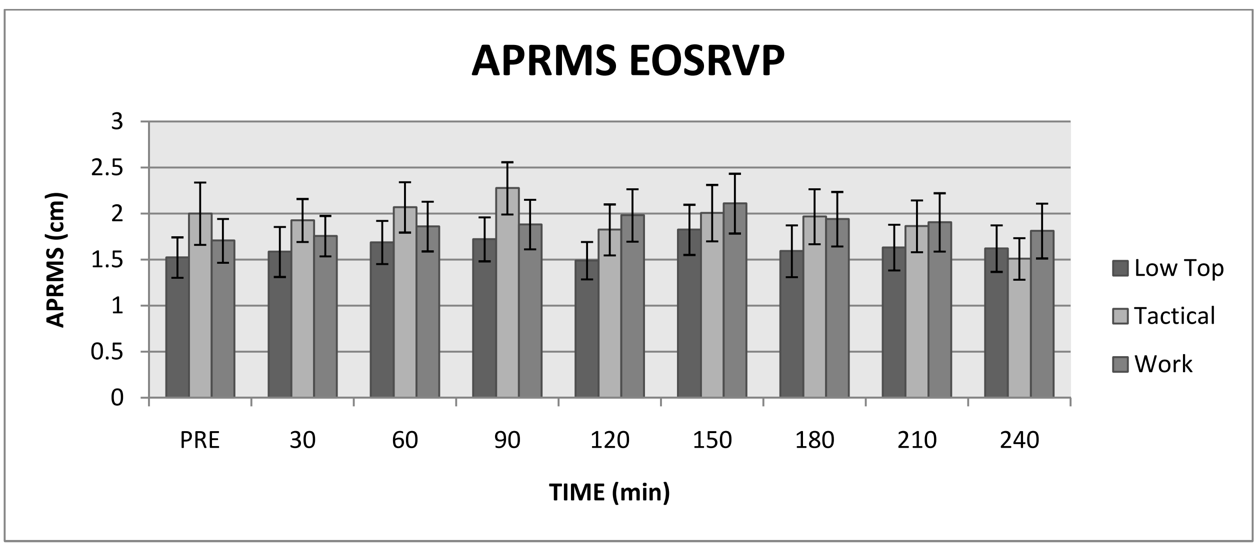


Figure 6:



<u>Anterior-Posterior Sway Velocity (APVEL)</u>

A repeated measures ANOVA with a Greenhouse-Geisser correction determined that there was a statistically significant difference in the Eyes Closed Sway Referenced Platform condition ($P = 0.018$) ($F (8, 64) = 2.537$, $P < 0.05$) and in the Eyes Open Sway Referenced Vision and Platform condition ($P = 0.009$) ($F (3.593, 28.740) = 4.362$, $P < 0.05$). The Post-hoc tests did not reveal any significance between the different time points for both the Eyes Closed Sway Referenced Platform condition and the Eyes Open Sway Referenced Vision and Platform. There was no other significant difference determined between the time points for the rest of the conditions of the SOT with Time and Shoe as the main effect and in the Time-Shoe Interaction.

Figures 7-12: Averaged Sway Velocity measures in the Anterior-Posterior directions for each of the six postural stability testing conditions. # indicates a significant difference over time intervals and * indicates a significant difference between the boot types and the bars represent the standard errors.

Figure 7:

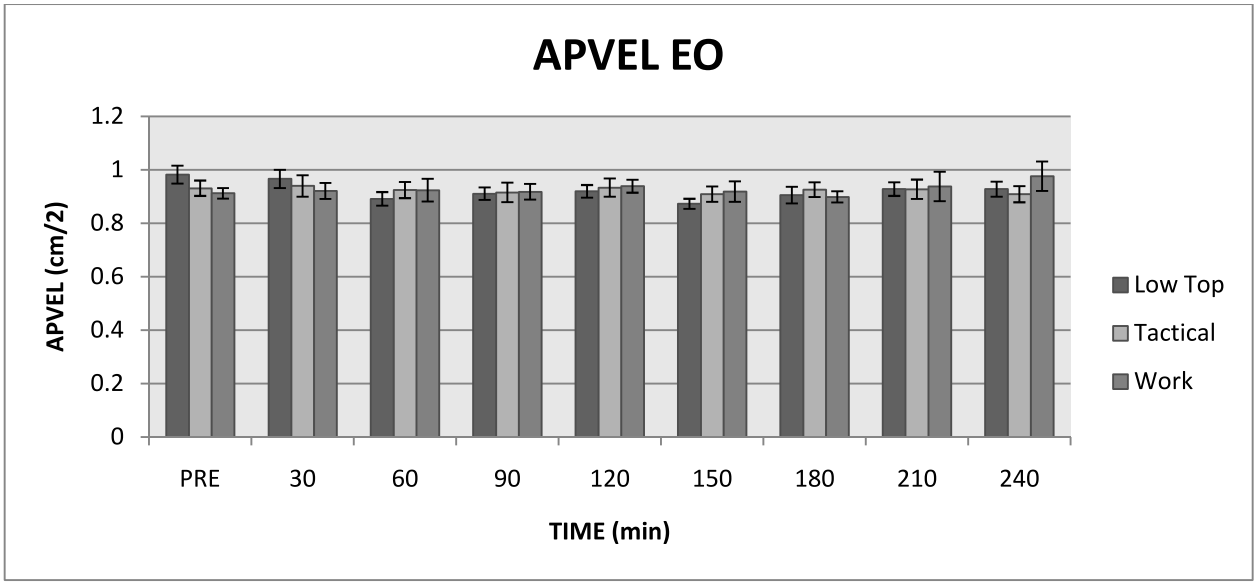
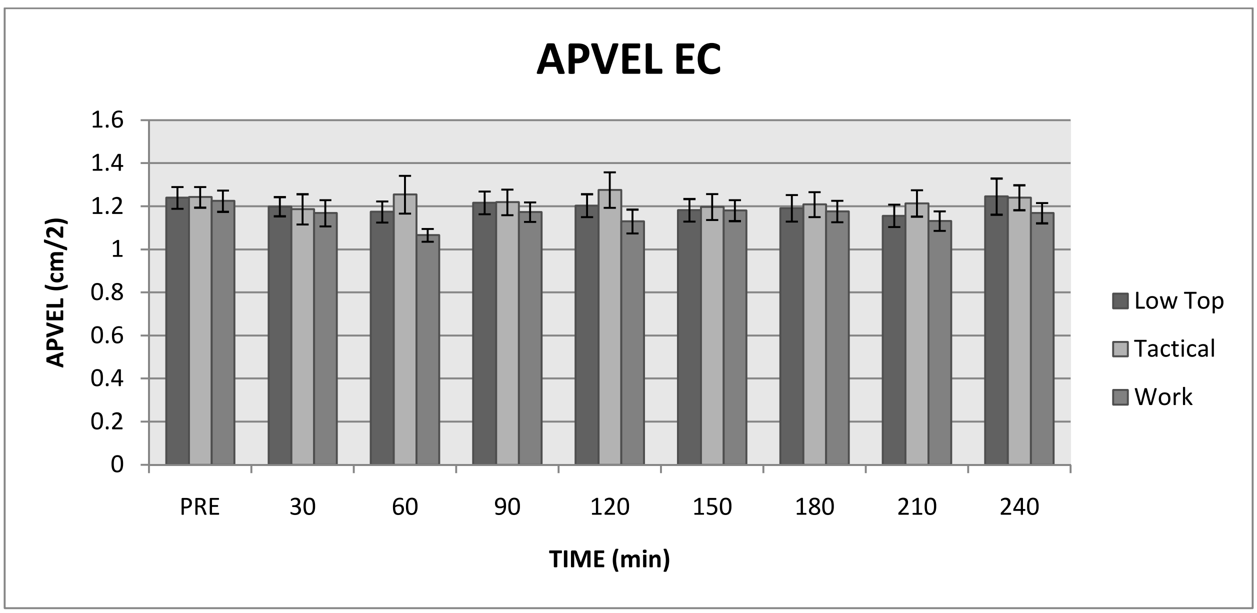


Figure 8:



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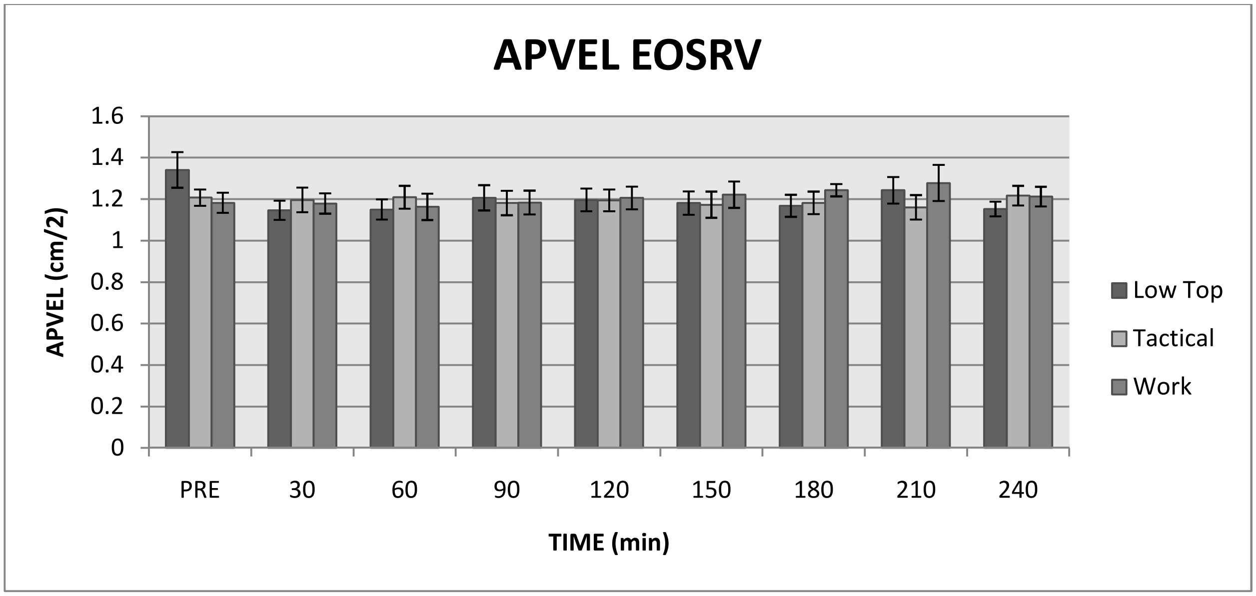

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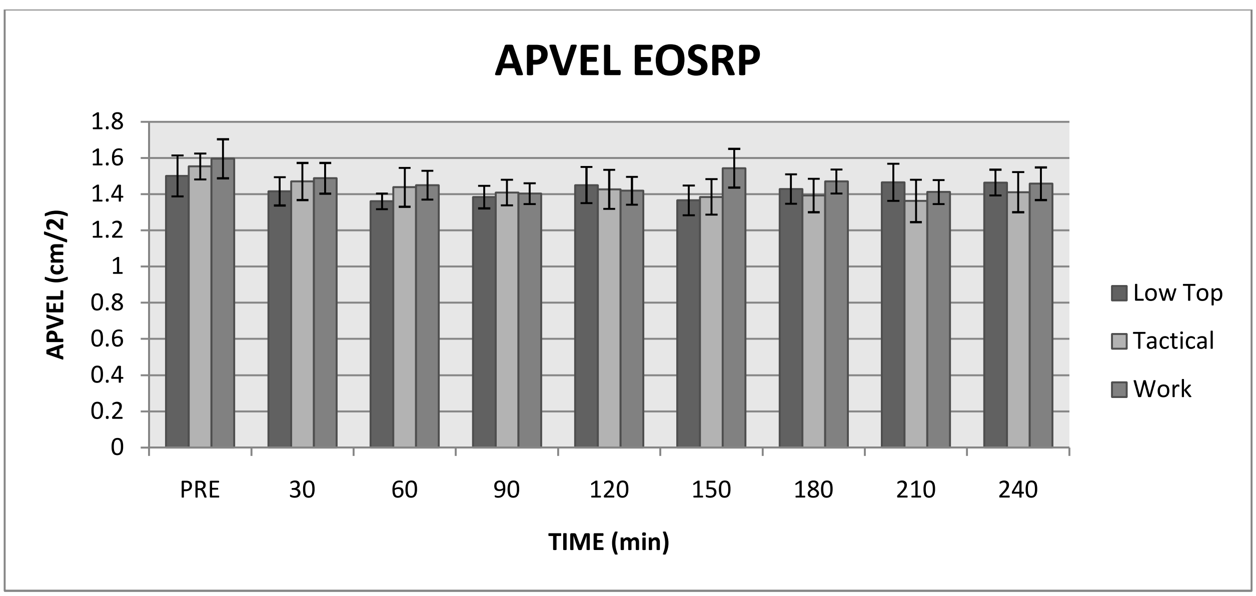

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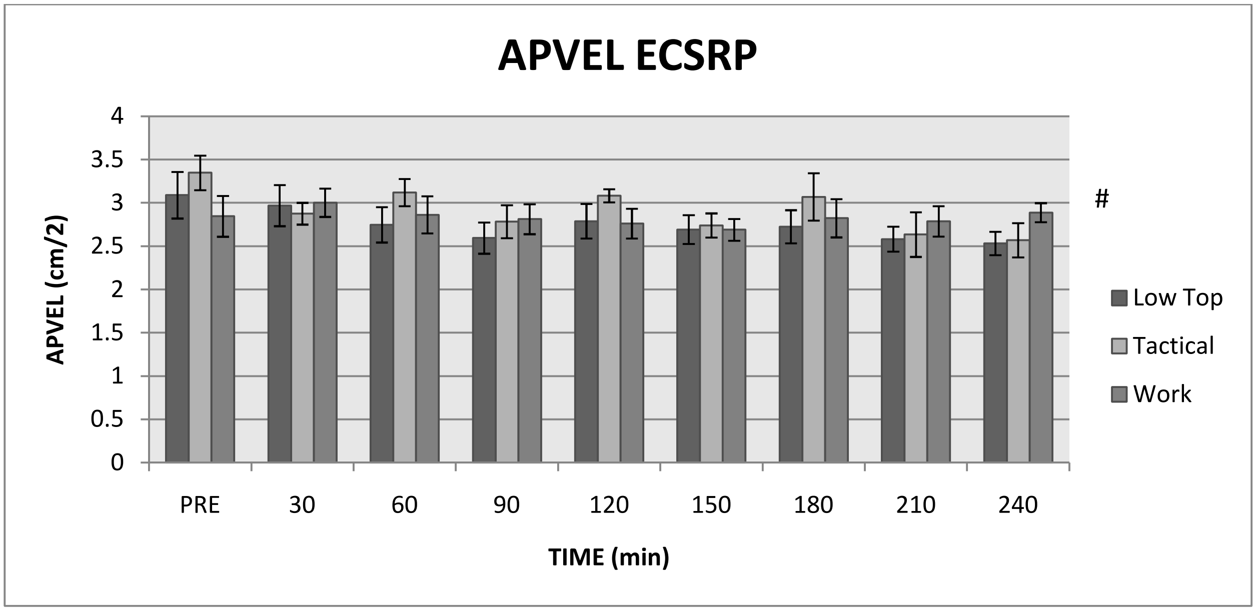
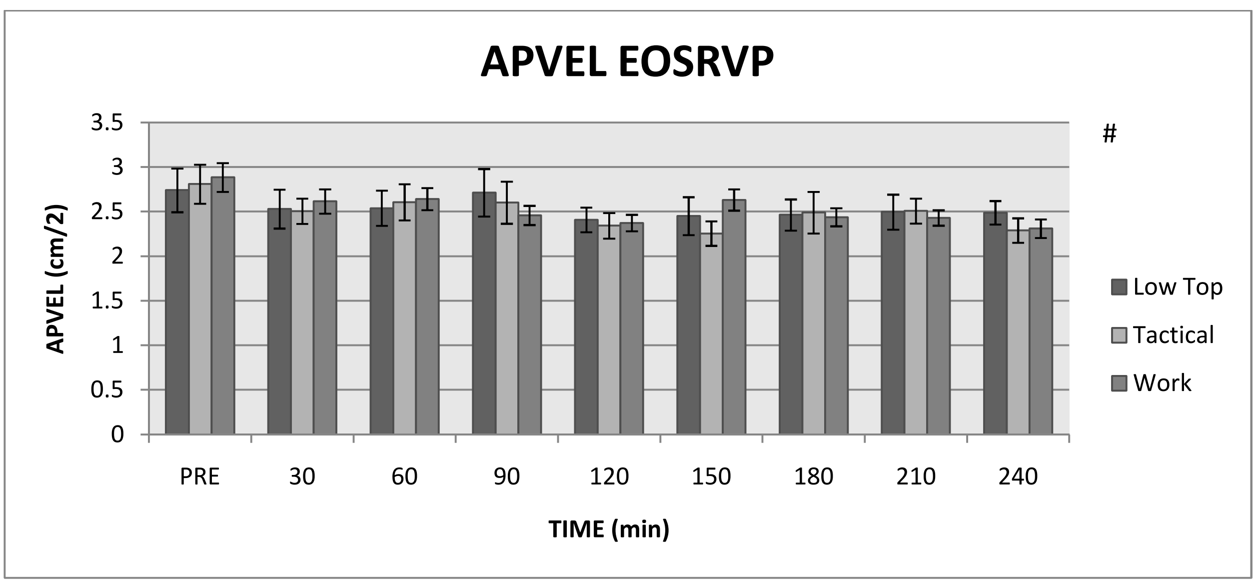

APVEL ECSRP
APVEL (cm/2)
4
3.5
3
2.5
2
1.5
1
0.5
0
PRE
30
60
90
120
150
180
210
240
TIME (min)
#
Low Top
Tactical
Work

Figure 12:


APVEL EOSRVP
APVEL (cm/2)
3.5
3
2.5
2
1.5
1
0.5
0
PRE
30
60
90
120
150
180
210
240
TIME (min)
#
Low Top
Tactical
Work

Medio-Lateral Sway RMS (MLRMS)

A repeated measures ANOVA with a Greenhouse-Geisser correction determined that there were statistically significant differences with time as the main effect in the Eyes Open ($P = 0.025$) ($F(8,88) = 2.347$, $P < 0.05$), Eye Closed ($P = 0.001$) ($F(8,72) = 3.786$, $P < 0.05$), Eyes Open Sway Referenced Vision ($P = 0.015$) ($F(8,72) = 2.602$, $P < 0.05$) and in the Eyes Open Sway Referenced Platform ($P = 0.006$) ($F(8,80) = 2.953$, $P < 0.05$) conditions of the SOT. The Post-hoc tests did not reveal any significance between the different time points for all the above mentioned conditions. There were no significant differences in the time-shoe interaction for all six conditions. However, significant differences existed in the Eyes Closed Condition ($P = 0.003$) ($F(2, 18) = .032$, $P < 0.05$) and in the Eyes Open Sway Referenced Platform Condition ($P = 0.006$) ($F(2, 20) = 9.959$, $P < 0.05$) with shoe type as the main effect. Post hoc tests using the Bonferroni correction revealed that the LB had a significantly greater sway RMS in the medio-lateral direction when compared to the TB and WB in the Eyes Closed Condition and had a significantly greater sway RMS in the medio-lateral direction when compared to both TB and WB in the Eyes Open Sway Referenced Platform condition. There was no statistically significant difference among the TB and WB in the above mentioned conditions.

Figures 13-18: Averaged Sway RMS measures in the Medio-Lateral directions for each of the six postural stability testing conditions. # indicates a significant difference over time intervals and * indicates a significant difference between the boot types and the bars represent the standard errors.

Figure 13:

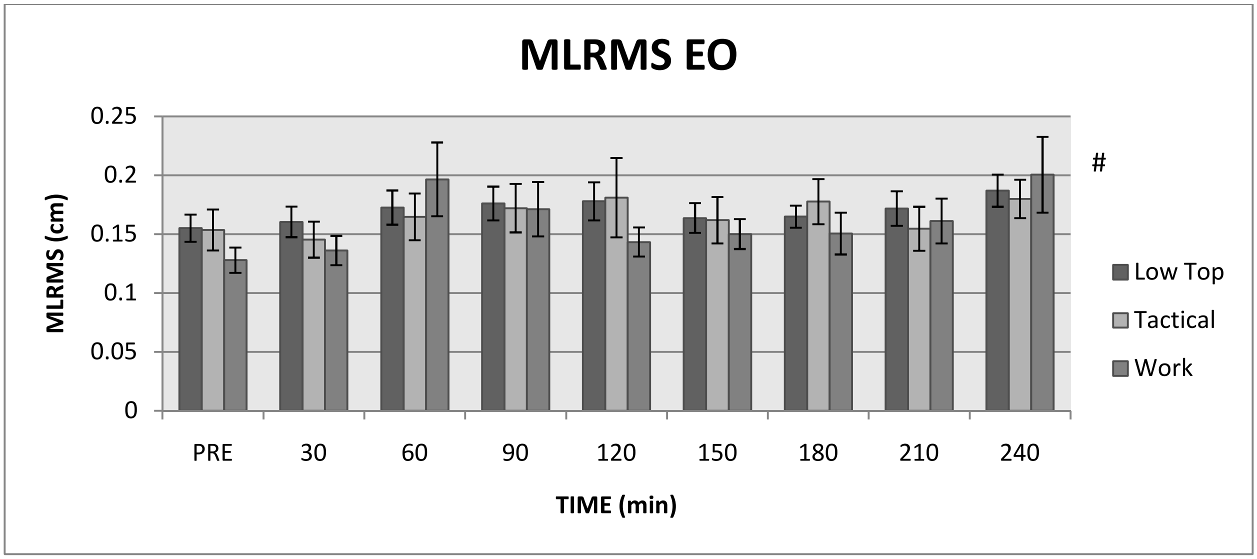
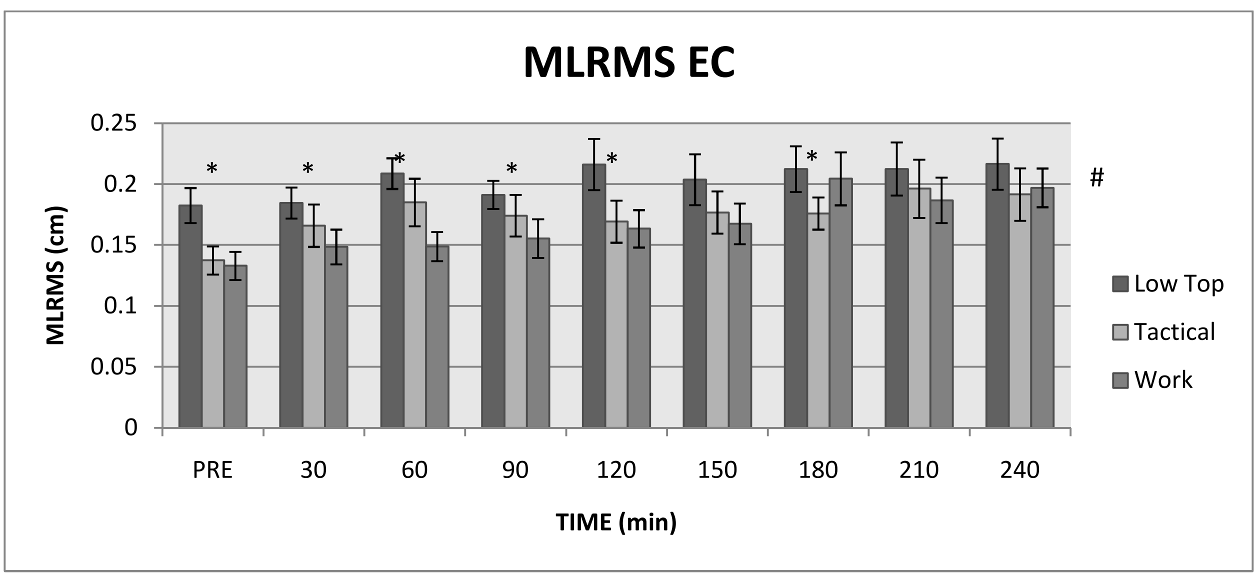


Figure 14:



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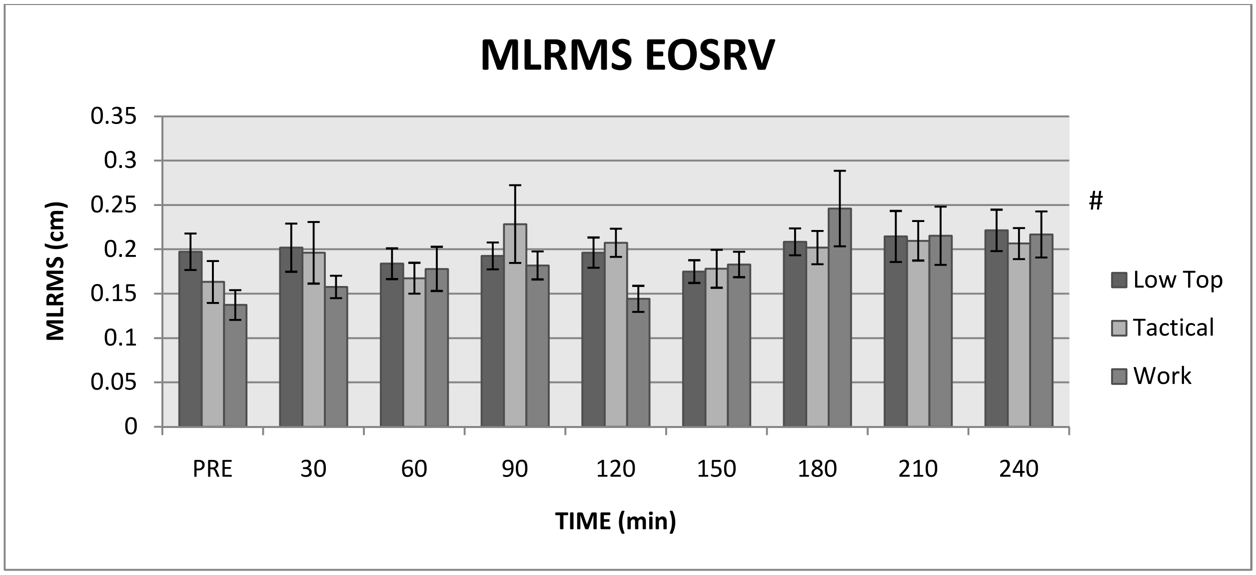

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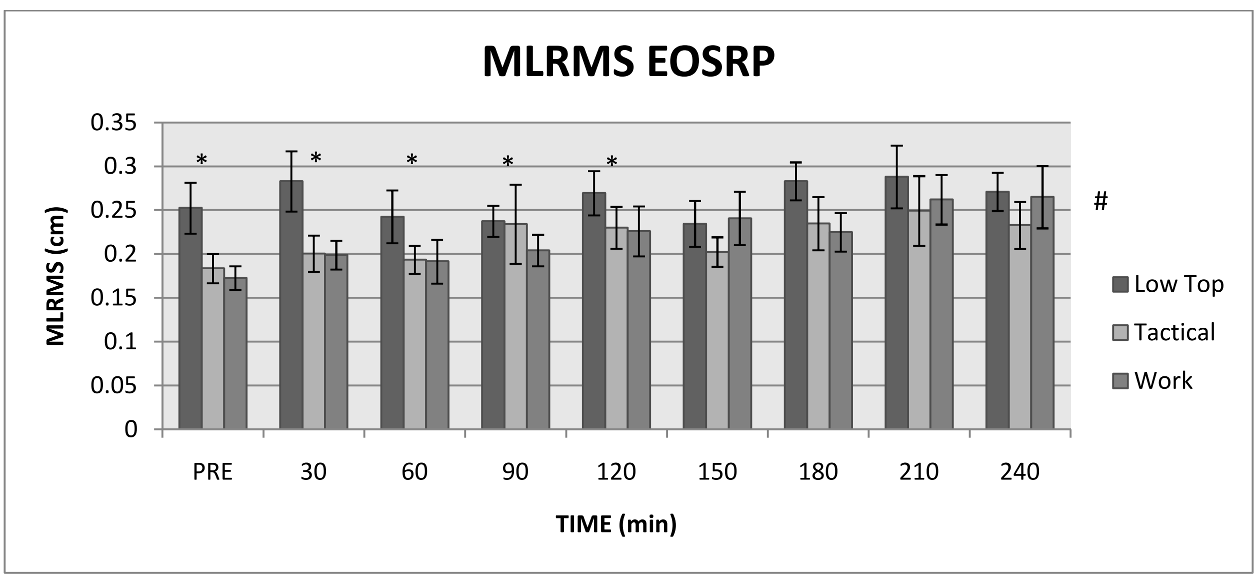

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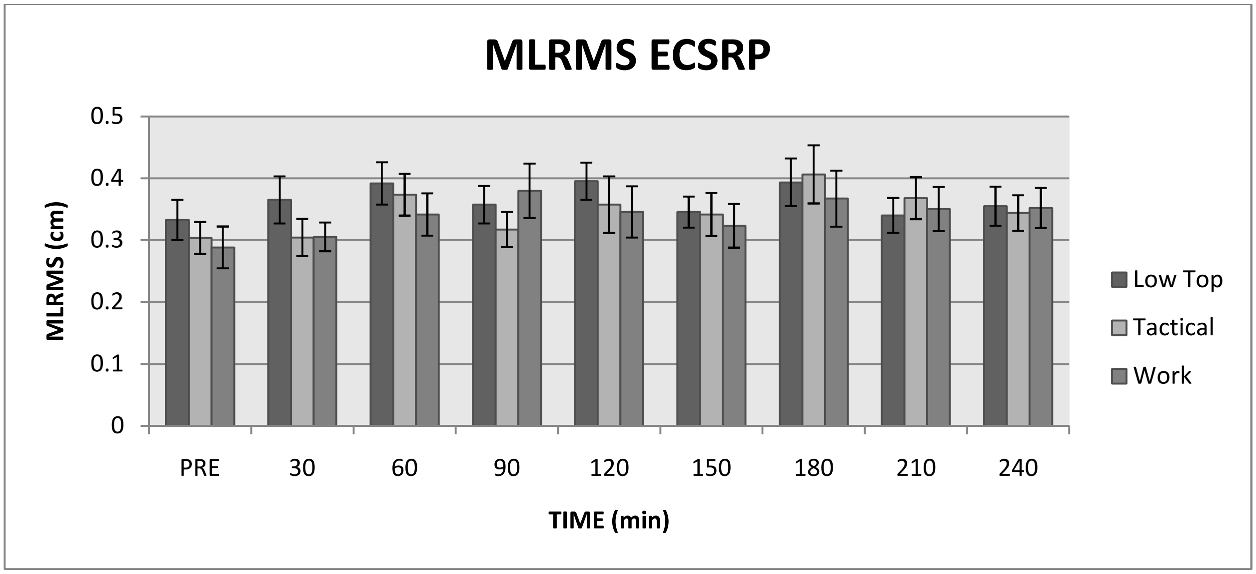

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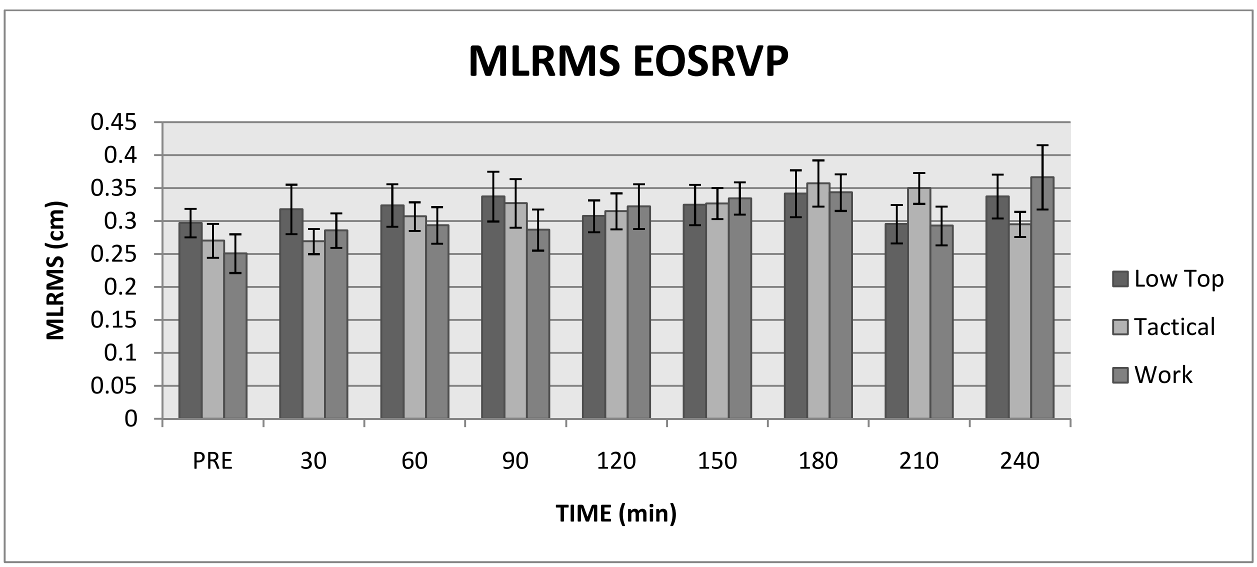
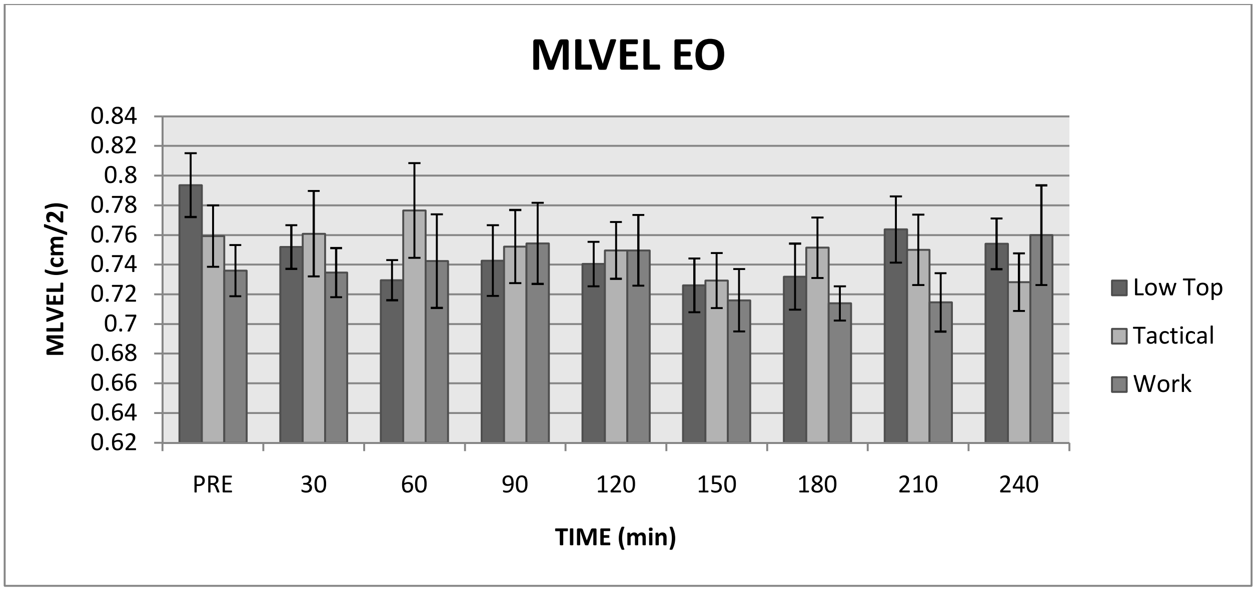


Medio-Lateral Sway Velocity (MLVELO)

A repeated measures ANOVA with a Greenhouse-Geisser correction showed that there were no significant differences determined across the time points for all of the conditions of the SOT with Time and Shoe as the main effect and no significant difference in the Time-Shoe Interaction.

Figures 19-24: Averaged Sway Velocity measures in the Medio-Lateral directions for each of the six postural stability testing conditions. # indicates a significant difference over time intervals and * indicates a significant difference between the boot types and the bars represent the standard errors.

Figure 19:



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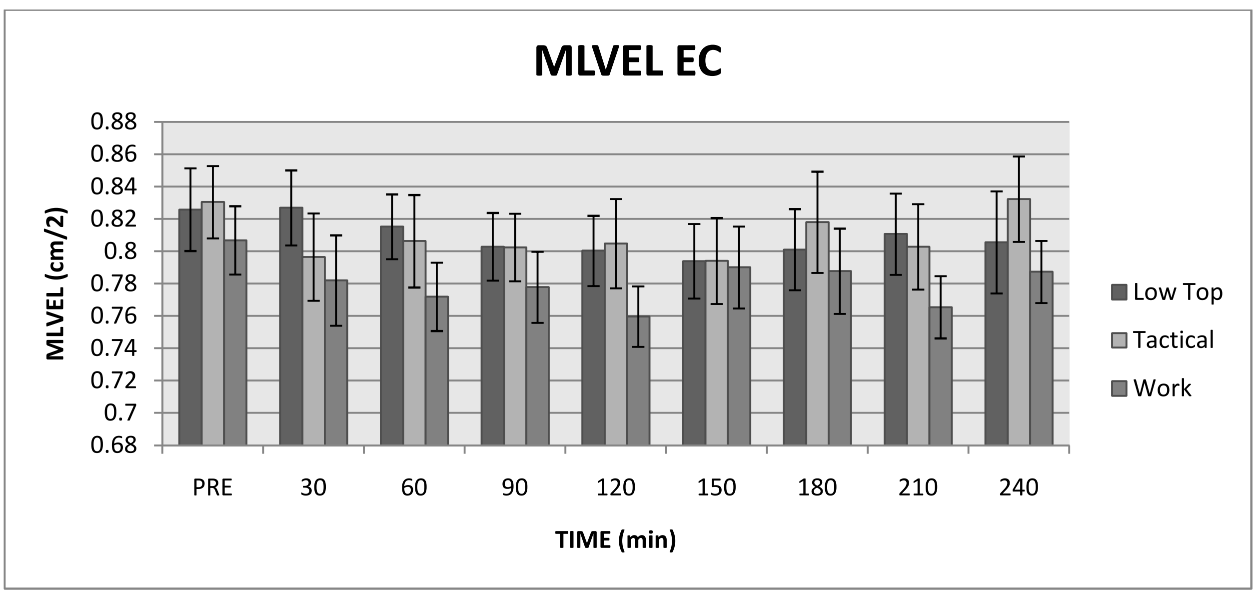
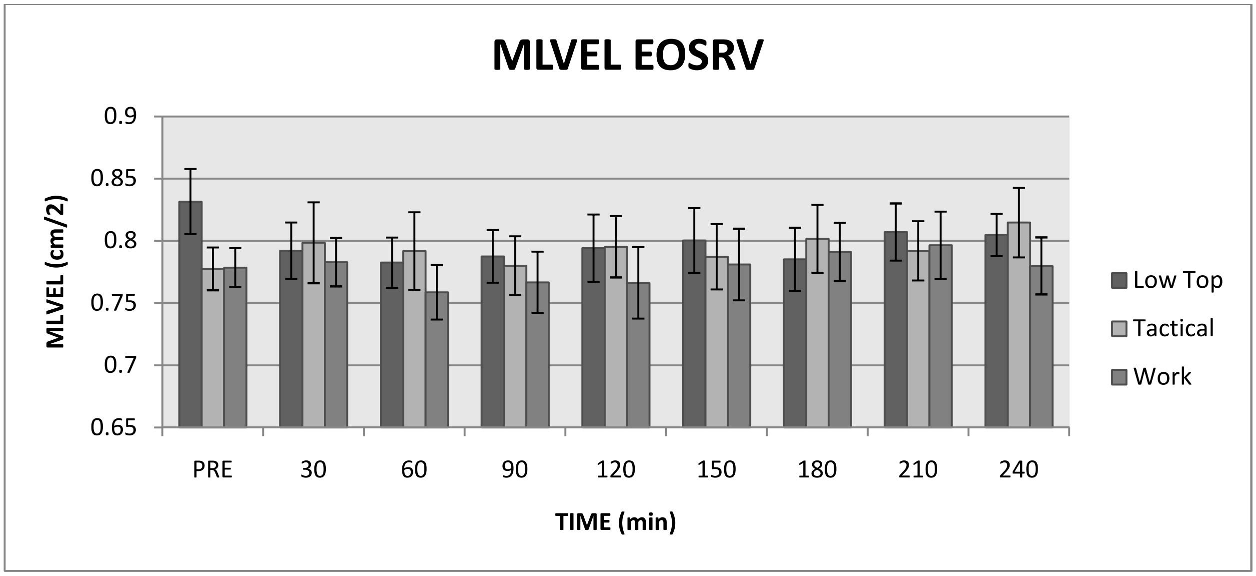


Figure 21:



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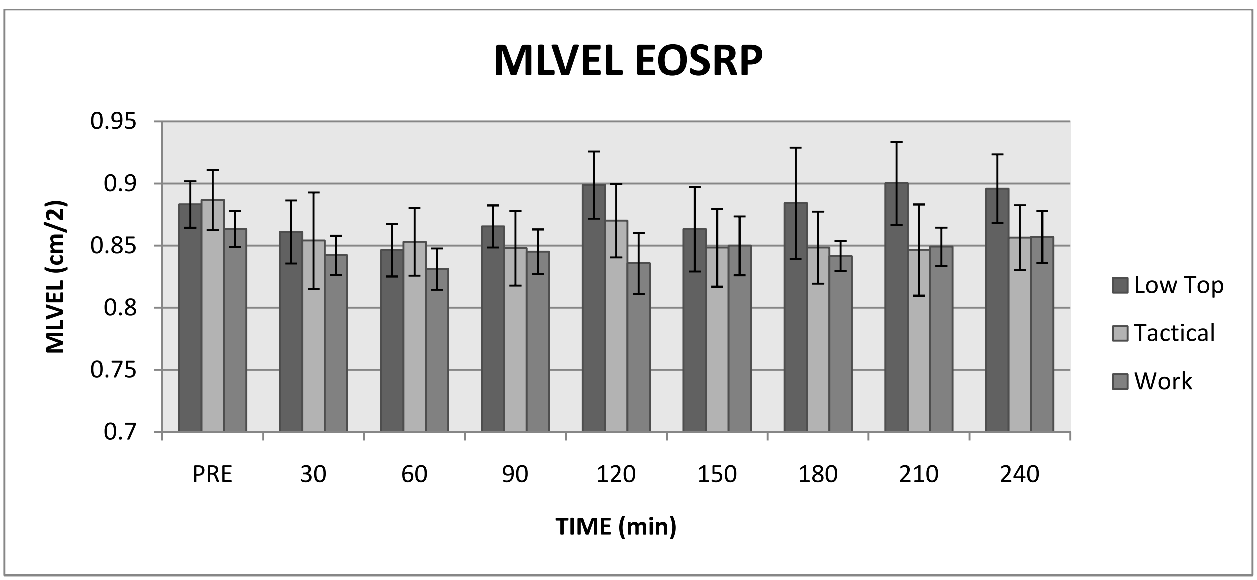


Figure 23:

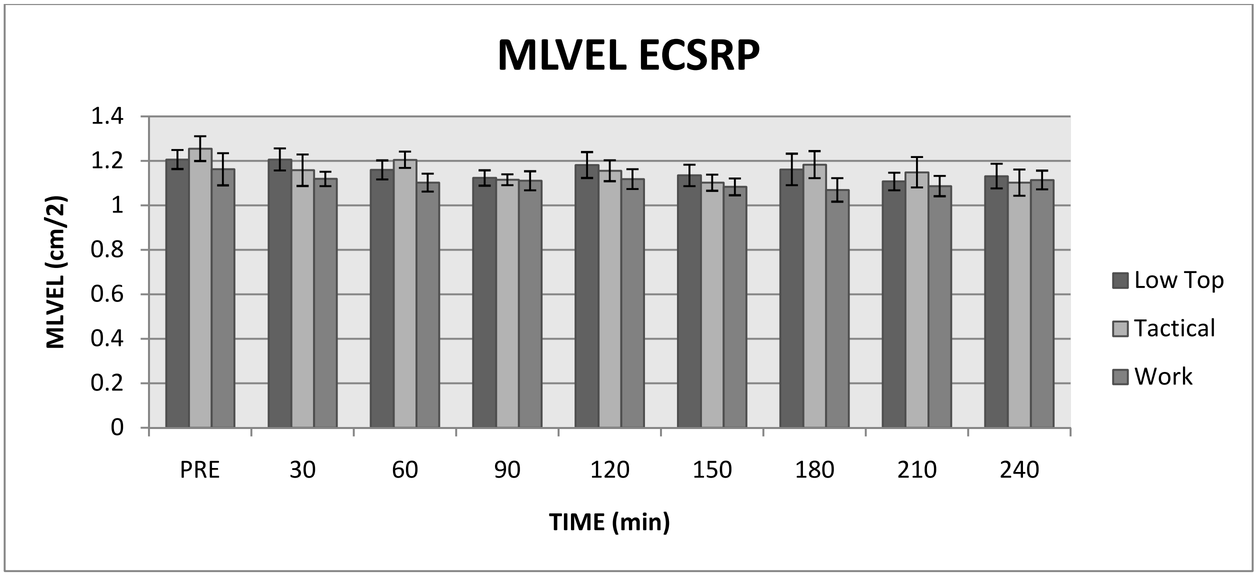
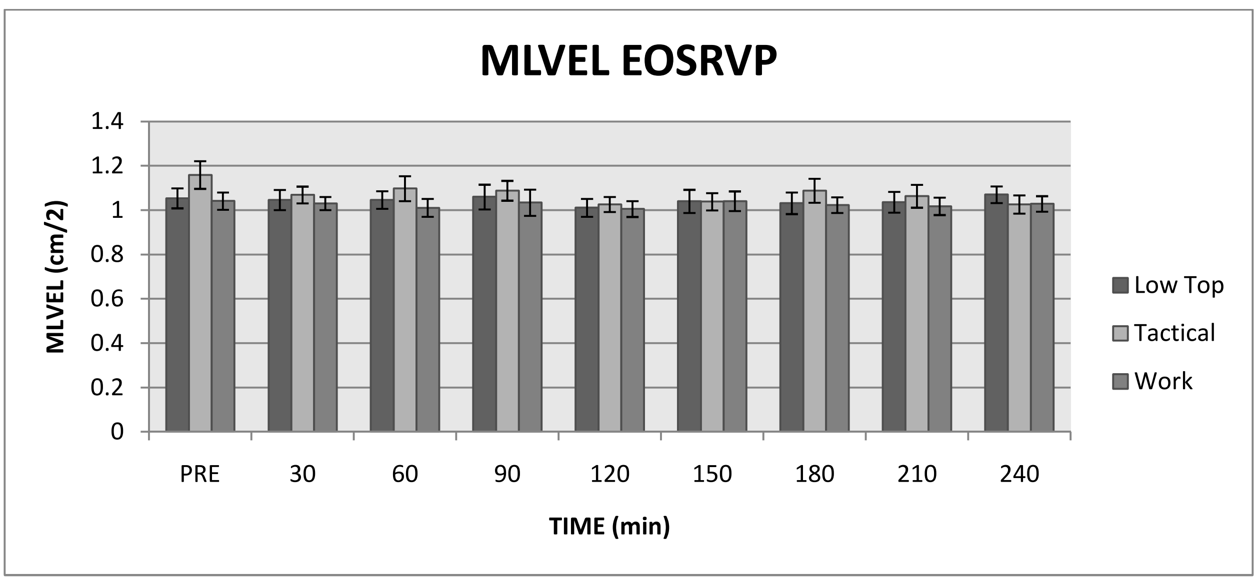


Figure 24:



CHAPTER V

DISCUSSION

This study examined the differences in balance among 14 healthy male adults wearing three types of footwear commonly used in the occupational industry which included the low-top slip resistant shoe (LT), tactical boot (TB) and work boot (WB), in response to a prolonged exposure to walking / standing on a hard firm surface. Balance was assessed by the outcome variables of anterior-posterior and medio-lateral sway velocities (APVEL and MLVEL) and the root mean square of the postural sway in the anterior-posterior and medio-lateral directions (APRMS and MLRMS). The objective of the present study was to analyze the mechanism of how postural control and balance differ in response to a work load which is induced at a slow rate over an extended period of time and how they differ among the three different types of footwear. There were significant differences found individually between footwear and time. No significant differences existed among the interaction between time and footwear. The discussion for this study is addressed with two major predictors of balance (i) balance performance between footwear types and (ii) balance performance over time.

Balance performance between footwear types:

The results of this study found significant differences in postural sway between the types of footwear. The significant differences occurred in the APRMS-SOT condition of eyes closed with no sway referencing of the surround and the support/platform (APRMS EC), in the MLRMS-SOT condition of eyes closed with no sway referencing of the surround and the support/platform (MLRMS EC) and in the eyes open sway referenced support/platform condition (MLRMS EOSRP). Pairwise comparison between the footwear types revealed that the WB and TB were significantly different from LT and performed better with significantly lower mean MLRMS when compared to the LT. No significant differences existed between the WB and TB.

The anatomical constraints (foot geometry, body mass and its distribution, or segment length and height), physiological constraints (muscular strength, rate of muscle force rise, or gains and delays of feedback control) and cognitive and behavioral constraints (reaction time, attention or fear of falling) play a major part in postural control (Redfern and Cham, 2001). In addition to the anatomical factors and foot geometry; footwear considered as the interface between the foot and the walking/standing surface, is of vital importance to human balance and postural control. The effectiveness of such footwear is responsible for a better and efficient transformation of the mechanical power output produced by the musculoskeletal (Cikajlo & Matjacic, 2007).

In addition to the protection and safety factors that are mandatorily required in the industrial foot wear, the performance aspect of such footwear has become an important requirement. The footwear used in this study were predominantly designed giving preference to the safety measures that are essential in an industrial or an occupational setting. Thus, these occupational footwear may fail to provide appropriate biomechanics that are needed for a normal gait and maintenance of balance by concentrating on the safety measures. The WB, which are equipped with steel toes or metatarsal guards that provide toe protection from impact and compression injuries, oil resistant soles, and an elevated boot shaft height that extends above the ankle joint with distinct heels, met ANSI-Z41-1991 standards as per the OSHA regulations for footwear in safety and protection (Occupational Safety and Health Administration, U.S. Department of Labor). The LT and TB are commonly used in a varied population for which there is not a prescribed OSHA regulation (OSHA, Laws and Regulations, 1970). Each of these footwear's mechanical characteristic and design feature such as the boot shaft height, boot shaft stiffness, heel height, mid-sole hardness, sole design and the mass of the footwear are of vital importance in maintenance of balance and each play a specific role in postural stability.

Boot shaft height as a predictor of balance performance:

Boot shaft height has been shown to be an important predictor for balance maintenance. A majority of the literature supports the general notion that a boot shaft elevated above the ankle joint increases the support around the ankle and offers greater postural stability (Cikajlo & Matjacic, 2007; Bohm & Hosl, 2010). The WB had the greatest shaft height of 18.5cm, followed by the TB with a shaft height of 16.5cm and the LT with a shaft height of 9.5cm. It has been shown that a high or an elevated boot shaft improves balance performance by providing support and stability to the ankle joint.. It has also been shown that circumferential pressure around the ankle enhances joint position sense and improves stability in a population with poor proprioception (You SH, Granata KP, Bunker LK 2004; Navrag B. Singh, Maury A. Nussbaum, Michael A. Madigan). The elevated boot shaft acts to provide compression around the ankle and serves as a stability measure at the ankle. Previous literature has shown that greater compression at the ankle may improve balance by increasing feedback from the cutaneous receptors in the foot and ankle and resulting in an improved joint position sense (Feuerbach JW, Grabiner MD, Koh TJ, Weiker GG 1994). The biomechanical function of the boot shaft is to restrict excessive inversion and thereby protect the ankle from very commonly encountered sprains such as the lateral collateral ligament sprain of the ankle. A considerable boot shaft thickness is required to perform this function effectively (Bohm & Hosl, 2010). The mechanical characteristics of the elevated boot shaft contribute in maintaining balance by providing support at the ankle. In all the shoes and boots where the boot shaft is higher than the ankle joint, the range of motion of the ankle joint is restricted (Cikajlo & Matjacic, 2007). This, in turn leads to alteration in the power generation at the ankle joint for propulsion during gait. Thus, boot shaft thickness has been shown to considerably affect and influence the kinematics and kinetics of the ankle, while a softer boot shaft allowed for greater range of motion and a greater generation of power during propulsion Cikajlo & Matjacic, 2007). A reduced power generation at the ankle joint can impair the stability during gait and during static stance. This may influence compensatory changes at the knee and at the hip joints. (Bohm & Hosl, 2010). Hence, it may also force a hip strategy for a fast recovery during a perturbation by using the hip muscles with a larger cross-sectional area. Although, the ankle strategy is attributed to the maintenance of postural stability by using small fine movements, the forced use of the hip strategy, which involves large gross movements, may be a better choice to recover because of the use of large muscles groups. The disadvantage of using the hip strategy is that the postural corrections in response to a perturbation may involve repeated corrections of over shooting the target of maintaining the CoG within the BoS with each attempt, which results in an increased postural sway. The hip muscles have a larger cross sectional area and have the ability to produce a greater amount of force. But, by the use of the proximal muscles to create compensatory contractions in the maintenance of balance, the force produced often translates into production of gross movement, which may induce greater postural sway (Gribble & Hertel, 2004). But, with extended durations of walking and with the possibility of localized muscular fatigue setting in, the elevated boot shaft might allow for a lesser workload on the ankle musculature and thereby limiting fatigue and providing better stability over time. These findings are in support of the current study's results in showing that an elevated boot shaft help maintain better balance in the WB and TB in comparison to the LT.

Heel height as a predictor of balance performance:

Heel height of the footwear is another important factor contributing to balance maintenance (Snow RE, Williams KR, 1994; Menant et al. 2008). A few explanations exist for the decrement in balance with the use of shoes with elevated heels. An anterior shift in the total center of mass of the body result in a modified posture and plantar pressure distribution and a smaller tipping angle compared to low top shoes. This results in lateral instability and may contribute to the decrement in balance with the use of shoes with elevated heels (Snow RE, Williams KR, 1994; Snow RE, Williams KR, Holmes GB Jr, 1992; Tencer AF et.al, 2004). The findings from Menant's footwear and balance study reported and confirmed that an elevated heel of 4.5 cm significantly impairs balance, whereas a hard shoe sole and a high heel collar may enhance balance in older people (Menant et al. 2008). Their findings also showed that an increased shoe heel height and sole softness caused a more conservative walking pattern and impaired ML balance control, respectively, in both young and older subjects (Menant et al. 2008). This is contradictory to this study's findings in which the WB with the highest heel height and the TB with the second highest were shown to perform better than the LT in which the heel height was the least. A couple of explanations can be drawn from this contrary finding. The effectiveness of the height of the boot shaft might have compensated for the elevated heel for the WB and the TB, leading to a better balance performance. Also, the WB and TB had a greater sole surface area, than that of the LT, which increases the BoS in normal stance. The width of the sole of the WB and TB were 9.6cm and 8.8cm, respectively in comparison to the 8.5cm for the

LT. The width at the forefoot for WB and TB were 12cm and 11cm in comparison to the 10.5cm for the LT. Load distribution in normal stance showed that the heel carried 60%, the midfoot carried 8% and the forefoot carried about 28% of the weight bearing load, with the toes only minimally involved in the weight bearing process (Cavanagh, 1987). The load distribution is better when it is dispersed across a greater surface area of sole of the foot wear, as in the case of WB and TB. This larger sole surface area provides a larger BoS and help in aiding balance maintenance. Hence, these factors might have been the reason behind the better performance of the WB and TB in comparison to the LT, even though the heel height was higher in these footwear.

In another relevant study, Menant et al. (Menant et al. 2008) showed that elevated shoe heels elicited reduced walking velocity and also showed that an elevated heel has a profound detrimental effect on balance maintenance and postural control. But, it was also reported that high-collar shoes led to a greater double support time and a greater step width, which are important in maintaining better balance by accomplishing a greater time period spent in double stance, which is a more stable phase of gait cycle than the period of single support. This has an effect of lowering the CoG during gait, and allowing for a greater step width by accomplishing a greater BoS. Even though, the outcome variables of this study were gait parameters, it still supports our results in explaining that high collar shoes or shoes with an elevated boot shaft such as the TB and WB perform better in balance tasks.

Mass of the footwear as a predictor of balance:

The mass of the boot also plays an important part in balance maintenance. An increase in the mass of the boot has been shown to cause an increase in energy expenditure by 0.7% - 1.0%

of locomotion for each 100g increase in the weight of the footwear (Jones, Toner, Daniels & Knapik, 1984, Martin, 1984). It was shown in a previous study with firefighters that two commonly used footwear in the firefighting industry, leather and rubber boots were significantly different from each other in balance maintenance. Both shoes had very similar characteristic design with moderately elevated heel and an elevated boot shaft. The only difference was the mass of the boot. The rubber boots were found to cause more decrements in balance in comparison with the leather boots over fatiguing fire simulation activity. The authors suggested that the higher mass of the rubber boot might have caused more fatigue which might have been the reason for the increased postural sway and thereby a poor performance in balance (Chander, et al. 2010). A general notion associated with the current study is that the WB, which had the greatest mass, should cause more fatigue over time and lead to a worse performance in balance assessment and the LT which had the least mass, should induce less fatigue and lead to better balance performance. But, this theory was contradictory to the findings of this study. It was the LT which was found to have the worst balance performance than all types of footwear. It could be hypothesized that the reason for this might have been that other positive design characteristics, such as the elevated boot shaft height and a larger BoS compensated for the differences which may have been caused by the mass of the footwear.

Sole design as a predictor of balance:

A positive feature of the TB is that it’s sole of resembles that of an athletic shoe, which might aid in a better gait kinematics and proper foot biomechanics. The LT has a very flat sole with a minimum surface area of BoS in a normal stance. Even though the WB provides a larger surface area for the BoS in a normal stance, it essentially holds the foot in a state of plantar flexion with its elevated heels. Furthermore, the TB combines the useful design features of both the LT and the WB and incorporates into a single type of footwear. Thus the TB has the advantage of having relatively less mass; with both an athletic sole design and an elevated boot shaft, thus offers a greater ability to maintain balance and postural stability in the industrial and occupational setting where the workload is presented over an extended period of time.

Mid-sole stiffness as a predictor of balance:

Although, this study did not measure mid-sole hardness, the shoes used in this study had difference in the nature and firmness of the mid sole. The LT had a softer mid soles in comparison with TB and WB. It has been shown that soft soles, even though they help prevent pain by providing a cushioning effect, may decrease the detection of pressure changes at the soles, and may have a negative effect on balance (Robbins S, Gouw GJ, McClaran J. 1992). However, the firm and hard mid soles may improve cutaneous and proprioceptive feedback and thereby may improve balance maintenance (Hijimans 2007, Menant 2008).

Balance performance over time:

The results of this study found a significant increase of postural sway RMS in the ML direction over time. The significant differences occurred in the MLRMS-SOT conditions of eyes open (MLRMS EO) and eyes closed (MLRMS EC) with no sway referencing of the surround or the support/platform and in the eyes open conditions with sway referenced vision (EOSRV) and support/platform (EOSRP) with the means increasing between the pre test and the 240th minute. However, post-hoc analysis did not reveal any significant difference between the 9 different testing time points. There were significant increases in the means of ML-RMS in all of the above mentioned testing conditions. This increase in postural sway may be attributed to fatigue caused by the continuous walking/standing for 4 hour duration. The significant differences in the EO and EC condition can be related to the detrimental effects to the somatosensory system. The significant increases in the means of MLRMS in the EOSRV-SOT condition implies that balance performance goes down when there is inaccurate visual information over time. And the significant increases in the means of the MLRMS in the EOSRP-SOT condition imply that balance performance goes down when there is inaccurate somatosensory information over time. Increases in postural sway and therefore detrimental effects to balance following prolonged duration workload have been supported by previous literature (Cham & Redfern, 2001; Wade & Davis, 2009; Lepers et al. 1994). The finding from this study directly support these literature with an increase in postural sway following an extended duration of walking / standing.

Previous literature, in which postural sway was found to increase due to fatigue supports this study's findings that balance is compromised with an increase in postural sway (Yaggie & McGregor, Lundin et al., Grible & Hertel). Their findings specifically support this study's findings of an increased postural sway in the ML direction which can be attributed to a lateral ankle instability resulting from fatigue. Lepers et al. showed that the ability to maintain postural stability during conflicting sensory conditions decreased after a prolonged exercise protocol, which is in direct support of this study's findings that a significant increase in ML RMS was found under conflicting sensory conditions of the SOT for balance assessment. Although, maintaining balance exclusively through an ankle strategy is achieved by the body functioning as an inverted pendulum, where there is relatively more sway at the distal end (head) from the axis of rotation at the ankle. But, in a hip strategy the knee and hip motions have been shown to predominate in maintaining balance. There are usually increased trunk and hip accelerations in maintaining balance while using the hip strategy predominately (Adlerton & Moritz, 2003).

Increases in postural sway have been associated with fatigue resulting from an increased workload in industrial settings. These increases in postural sway in turn, are responsible for greater potential risk for encountering a fall or a slip (Parijat & Lockhart, 2008). Postural sway is essentially a result of continuous corrections and over corrections of lower extremity joint movements in an attempt to keep the body's center of gravity within the base of support. Since fatigue slows down the neural transmission, the ability to effectively compensate for the corrections about the joint movements is reduced (Gribble & Hertel, 2004). Hence, this correction over-correction cycle results in greater amplitude, increasing the sway amplitude and velocity. An increased postural sway was shown when firefighters were tested under fatiguing conditions while donning the personal protective equipment (Kincl et al. 2002). A similar study assessed the impact of working long shifts wearing the turnout gear and self-contained breathing apparatus on postural stability of firefighters. Postural stability was shown to decrease as the firefighters spent more time on duty (Sobeih et al. 2006). Increasing fatigue level can compromise multiple aspects of the neuromuscular system which are responsible for postural control and balance maintenance.

One of the major factors for the onset of fatigue is the time over which it is induced. Pline et al. (Pline et al. 2006) asserted that, inside the laboratory, localized muscle fatigue is typically induced at a higher workload over a period of several minutes, whereas, outside the laboratory, the localized muscular fatigue is frequently induced at a relatively low workload over a period of several hours. Pline at al. (Pline et al. 2006) reported larger increases in sway velocity and sway area when fatigue was induced over longer durations and at higher fatigue levels which supports the results of this study. Few studies have shown that muscular fatigue which is responsible for a decrement in balance performance is only short lived (Alderton et.al. 2003, Nardone et.al. 1997, Yaggie & McGregor, 2002). But, in all these studies, fatigue was induced over a short period of time by through the performance of specific fatigue-causing exercise protocols, unlike this study, which focused on reciprocating an industrial work setting with a relatively low workload (standing/walking) over an extended period of time.

Perturbations caused by the muscle fatigue in regard to joint position sense have been related to decreased motor neuron output (Yaggie & McGregor, 2002). Various fatigue models have been suggested with regard to different levels of the nervous system. At the peripheral level, a failure of the muscle to respond to a neural signal or a failure of the muscle to respond to neural excitation can be attributed to muscular fatigue. At the central level, fatigue is known to result in a failure of excitation of the motor neurons, which is caused by changes in the nervous system. The changes in the motor neuron firing have been attributed to the intrinsic properties of the motor neurons, recurrent inhibition due to the renshaw cells and changes in reflex inhibition. Changes in the descending drive in the motor neuron pool could also be responsible for changes in the descending drive in the motor neurons (Corbeil, Blouin, Begin, Nougier & Teasdale, 2003). Impaired joint position sense and impaired excitation-contraction coupling due to the decreased calcium ion availability for release from the sarcoplasmic reticulum have also been suggested as possible mechanisms for fatigue. The intrafusal fibers which are responsible for modulating the sensitivity of the muscle spindles that relay sensory information on the muscle velocity and length via the group Ia and II afferents respectively. So, any disruption of these functions due to fatigue can interfere with the functions of the spindle which are important for maintaining balance (Pline, et.al 2006).

Small perturbations in quiet stance are usually minimized by the use of the ankle strategy and the stretch responses from the muscle spindle (Yaggie & McGregor, 2002). However, these strategies and responses as well as their ability to minimize small disturbances and perturbations are compromised when local muscular fatigue is occurs. These strategies and responses would be expected to resume their normal function once recovery from fatigue occurs. This was explained by Yaggie & McGregor, as they proposed that as the recovery from fatigue progresses, the response from type 3 and 4 muscle afferents may have been increased, yielding an increase in somato-senssory sensation. Thus, this increase in proprioceptive input may have increased reflexive postural responses, resulting in better maintenance of posture (Yaggie & McGregor, 2002).

Contrary to the existing literature, postural sway velocity in the AP direction was found to decrease with time as the main effect in ECSRP and EOSRVP-SOT conditions. The ECSRP and EOSRVP SOT conditions are predominantly a measure of the use of vestibular system in maintenance of balance where inaccurate sensory information for the somatosensory systems with absent vision and inaccurate sensory information for both somatosensory and vision are presented by the SOT respectively. Possible explanations for this contrast finding may be attributed to the anticipatory or proactive postural responses. The cognitive system of postural control has both an adaptation and an anticipatory postural mechanism, which are reactive and proactive postural responses. Adaptation is seen when there is a decreased sway and amplitude of responses when a perturbation is given repeatedly over and over again, but anticipation is seen where responses are modified based on a central set pattern or our anticipation of the size and direction of the perturbation. This study's results reflect an anticipatory proactive postural mechanism during the conditions ECSRP and EOSRVP. Although the sway gain was set at +1.0 and the sway referencing of the visual surround and the platform do not occur unless a change in the participants CoP occurs, there could still be a psychological learning effect of standing in the same confined space of the Neurocom Equitest while assessing balance. Visual information is used to predict trajectory and estimate forces required for anticipated movement. The ECSRP and EOSRVP are both a test of predominantly the vestibular system in an eyes closed condition and with conflicting visual and sensory information, where visual preference of each individual assesses how they rely on the vision even when the visual information is incorrect. The improved sway velocity values might infer that the participants relied less on the visual information when it was incorrect with better use of the vestibular system. Hence, an anticipatory proactive postural response may be suggested for this contrary finding for the ECSRP and EOSRVP-SOT conditions.

The results from the paired sample t-tests revealed at which time period the significanct differences existed between the footwear types. Significant difference between WB, TB and LT for the APRMS EC condition was found at all time intervals except 30th minute of testing. This implies that without regard to the workload over the course of 4 hour duration, the differences between the footwear existed throughout the testing session. This finding may suggest that the WB and TB were successful in helping maintain a better balance performance with low mean RMS of postural sway in the AP direction at all time points except at the 30th minute of testing. The differences for the MLRMS EC condition between the WB, TB and LT occurred at all the testing time points except 150th, 210th and 240th minute of testing. The differences in footwear for the MLRMS EOSRP condition had similar results with significant differences occurring between WB, TB and LT until 120th minute and losing significance after the 150th minute of testing. This implies that the WB and TB aided in better performance in balance in comparison to the LT until the 150th minute after which there were no significant differences between the boot types. This finding can be related to the Cham and Redfern's in which it was shown that significant differences in CoP weigth shifts did not occur until at least the 3rd hour of a 4 hour testing session. The increase in discomfort and fatigue were found to be related to the increases in the CoP shifts which were significantly affected by the flooring types, but was evident only after the 3rd hour of standing (Cham & Redfern, 2001). In the current study the flooring type was kept constant and the interface between the foot and floor, the footwear was manipulated. Hence, Cham and Redfern's results (Cham & Redfern, 2001) could be related to this study's findings in suggesting that it was after the 150th minute that the significant differences were lost between the footwear types. This loss of significant differences between the footwear may be due to fatigue setting in after the 150th minute.

Conclusions:

Slips and fall-related injuries have been identified as a significant burden to the industrial working population. It is reported that floors, walkways or ground surfaces are a major extrinsic factor contributing to 86% of slip and fall related accidents. Occupationally induced muscle fatigue has been identified as a major intrinsic factor to induce falls and slips (Prakriti Parijat & Thurmon E Lockhart, 2008). Footwear, because it serves as the interface between the foot and the floor, is of vital importance to maintain balance and postural stability in the industrial and occupational settings and is a more appropriate and a simple choice of modification to ensure better static and dynamic stability.

The findings from this study can be used as series of recommendations for future occupational footwear design with regard to boot characteristics and their functions. Occupational and industrial footwear must serve multiple functions, aiding in both the safety and the appropriate biomechanics of the foot in static and dynamic stability. Hence, the following recommendations can be made from the findings of this study for efficient future designing of occupational and industrial footwear. Footwear with an elevated boot shaft which extends above the ankle joint, with firm midsole and an athletic outsole with relatively lesser mass will help in addressing the important mechanical characteristics of the footwear, which can aid in better balance performance. The results of this study may also help explain the decrements in balance with an extended period of workload involving continuous walking/standing that are most commonly seen in the occupational and industrial settings. Thus, a better understanding of balance with extended duration exposure in occupational footwear may help aid to minimize and reduce the number of fall related injuries in the occupational and industrial settings.

Future Research:

This study did not account for the hardness or the firmness of the midsole; the coefficient of friction of the footwear's slip resistant capabilities, shock attenuation capabilities of the footwear inserts, or the availability of rear foot motion during dynamic stability. Future researches on these mechanical characteristics of the footwear are recommended to have a better understanding of the footwear functions and their importance in postural stability. An in depth analysis of gait events in these footwear, along with joint kinetics, kinematics and muscle activity may be recommended.

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LIST OF APPENDICES

APPENDIX: A

DESCRIPTIVE STATISTICS

Table 3: Descriptive Statistics – APRMS – EO

	Descriptive Statistics - APRMS EO - CONDITION 1 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.3511	0.1001	0.0289	0.2960	0.0781	0.0225	0.2712	0.0488	0.0141
30	0.3268	0.0725	0.0209	0.2986	0.0873	0.0252	0.2812	0.0584	0.0169
60	0.3059	0.0515	0.0149	0.3213	0.1242	0.0359	0.3115	0.0755	0.0218
90	0.3175	0.0896	0.0259	0.2859	0.0744	0.0215	0.3294	0.0879	0.0254
120	0.3129	0.0642	0.0185	0.2880	0.0846	0.0244	0.3281	0.0658	0.0190
150	0.2910	0.0539	0.0156	0.2588	0.0658	0.0190	0.2990	0.0810	0.0234
180	0.3569	0.1135	0.0328	0.3013	0.0873	0.0252	0.2901	0.0753	0.0217
210	0.3048	0.0713	0.0206	0.3110	0.1063	0.0307	0.3133	0.0961	0.0277
240	0.3144	0.0679	0.0196	0.2722	0.0455	0.0131	0.3427	0.1027	0.0297

Table 4: Descriptive Statistics – APRMS – EC

	Descriptive Statistics - APRMS EC - CONDITION 2 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.5003	0.1377	0.0397	0.4287	0.1046	0.0302	0.4321	0.1125	0.0325
30	0.4444	0.1221	0.0353	0.3983	0.0699	0.0202	0.4300	0.0908	0.0262
60	0.5085	0.1366	0.0394	0.4503	0.1293	0.0373	0.4056	0.0670	0.0194
90	0.5025	0.1535	0.0443	0.4028	0.0665	0.0192	0.3940	0.0986	0.0284
120	0.4630	0.1098	0.0317	0.4614	0.1423	0.0411	0.4074	0.0876	0.0253
150	0.4778	0.1542	0.0445	0.4062	0.0969	0.0280	0.4279	0.1028	0.0297
180	0.4752	0.1236	0.0357	0.4133	0.1085	0.0313	0.4361	0.1093	0.0316
210	0.4703	0.1096	0.0316	0.4211	0.0964	0.0278	0.4101	0.0873	0.0252
240	0.5114	0.1197	0.0345	0.4213	0.0987	0.0285	0.4167	0.1012	0.0292

Table 5: Descriptive Statistics – APRMS – EOSRV

	Descriptive Statistics - APRMS EOSRV - CONDITION 3 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.5368	0.2398	0.0692	0.4413	0.1466	0.0423	0.4094	0.1017	0.0294
30	0.4199	0.1516	0.0438	0.4052	0.1373	0.0396	0.4405	0.1308	0.0378
60	0.4668	0.1793	0.0517	0.4653	0.1424	0.0411	0.4655	0.1926	0.0556
90	0.4654	0.1444	0.0417	0.4379	0.1598	0.0461	0.4098	0.1582	0.0457
120	0.4926	0.0959	0.0277	0.4406	0.1506	0.0435	0.4255	0.1049	0.0303
150	0.4646	0.1112	0.0321	0.4222	0.1156	0.0334	0.4517	0.1613	0.0466
180	0.4728	0.2080	0.0600	0.4052	0.1158	0.0334	0.4464	0.1111	0.0321
210	0.5189	0.1795	0.0518	0.4095	0.1622	0.0468	0.5062	0.2437	0.0703
240	0.4608	0.1455	0.0420	0.4170	0.0799	0.0231	0.4371	0.1556	0.0449

Table 6: Descriptive Statistics – APRMS – EOSRP

	Descriptive Statistics - APRMS EOSRP - CONDITION 4 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.8707	0.6202	0.1790	0.6762	0.2909	0.0840	0.8734	0.6021	0.1738
30	0.7305	0.3193	0.0922	0.7550	0.5158	0.1489	0.7824	0.4678	0.1350
60	0.7824	0.3832	0.1106	0.7249	0.4213	0.1216	0.8248	0.4215	0.1217
90	0.7497	0.3265	0.0943	0.8113	0.4543	0.1311	0.7586	0.4152	0.1199
120	0.8132	0.3637	0.1050	0.7315	0.3840	0.1108	0.9062	0.4550	0.1314
150	0.8152	0.4201	0.1213	0.7931	0.3338	0.0964	0.8885	0.6353	0.1834
180	0.8725	0.5580	0.1611	0.7350	0.3324	0.0960	0.8809	0.5091	0.1470
210	0.9202	0.5485	0.1583	0.7099	0.2761	0.0797	0.7916	0.3814	0.1101
240	0.8500	0.5918	0.1709	0.6936	0.3392	0.0979	0.8033	0.3995	0.1153

Table 7: Descriptive Statistics – APRMS – ECSRP

	Descriptive Statistics - APRMS ECSRP - CONDITION 5 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.5959	0.7212	0.2175	1.6431	0.6315	0.1904	1.5716	0.5963	0.1798
30	1.6352	0.7198	0.2170	1.6332	0.6254	0.1886	1.6343	0.5669	0.1709
60	1.6096	0.5996	0.1808	1.6846	0.5163	0.1557	1.7314	0.5481	0.1653
90	1.4659	0.4577	0.1380	1.5486	0.5951	0.1794	1.6924	0.6052	0.1825
120	1.6376	0.6566	0.1980	1.7162	0.7171	0.2162	1.7978	0.6998	0.2110
150	1.6306	0.7192	0.2169	1.5723	0.7419	0.2237	1.6869	0.7387	0.2227
180	1.6544	0.7819	0.2358	1.6696	0.6338	0.1911	1.8901	1.0136	0.3056
210	1.5957	0.5798	0.1748	1.5630	0.6934	0.2091	1.9206	0.9353	0.2820
240	1.7293	0.8199	0.2472	1.5136	0.6246	0.1883	1.7101	0.8091	0.2440

Table 8: Descriptive Statistics – APRMS – EOSRVP

	Descriptive Statistics - APRMS EOSRVP - CONDITION 6 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.5246	0.7611	0.2197	2.0002	1.1715	0.3382	1.7062	0.8244	0.2380
30	1.5848	0.9403	0.2715	1.9267	0.8095	0.2337	1.7573	0.7598	0.2193
60	1.6878	0.8102	0.2339	2.0692	0.9469	0.2733	1.8614	0.9367	0.2704
90	1.7225	0.8272	0.2388	2.2762	0.9803	0.2830	1.8824	0.9307	0.2687
120	1.4906	0.7041	0.2032	1.8249	0.9579	0.2765	1.9823	0.9831	0.2838
150	1.8256	0.9426	0.2721	2.0071	1.0618	0.3065	2.1100	1.1238	0.3244
180	1.5933	0.9718	0.2805	1.9677	1.0347	0.2987	1.9411	1.0246	0.2958
210	1.6328	0.8624	0.2489	1.8643	0.9752	0.2815	1.9057	1.0988	0.3172
240	1.6220	0.8743	0.2524	1.5096	0.7781	0.2246	1.8120	1.0293	0.2971

Table 9: Descriptive Statistics – APVEL – EO

	Descriptive Statistics - APVEL EO - CONDITION 1 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.9814	0.1156	0.0334	0.9307	0.0999	0.0288	0.9116	0.0684	0.0198
30	0.9655	0.1184	0.0342	0.9391	0.1376	0.0397	0.9206	0.1024	0.0296
60	0.8908	0.0888	0.0256	0.9237	0.1041	0.0300	0.9233	0.1467	0.0424
90	0.9103	0.0816	0.0236	0.9151	0.1261	0.0364	0.9175	0.1006	0.0290
120	0.9191	0.0814	0.0235	0.9329	0.1175	0.0339	0.9380	0.0831	0.0240
150	0.8722	0.0660	0.0191	0.9085	0.0990	0.0286	0.9183	0.1320	0.0381
180	0.9051	0.1087	0.0314	0.9251	0.0961	0.0277	0.8982	0.0735	0.0212
210	0.9273	0.0874	0.0252	0.9267	0.1260	0.0364	0.9373	0.1902	0.0549
240	0.9273	0.0978	0.0282	0.9083	0.1046	0.0302	0.9753	0.1910	0.0551

Table 10: Descriptive Statistics – APVEL – EC

	Descriptive Statistics - APVEL EC - CONDITION 2 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.2394	0.1683	0.0507	1.2424	0.1596	0.0481	1.2245	0.1640	0.0494
30	1.1985	0.1477	0.0445	1.1861	0.2336	0.0704	1.1680	0.2023	0.0610
60	1.1742	0.1621	0.0489	1.2543	0.2909	0.0877	1.0657	0.0969	0.0292
90	1.2166	0.1753	0.0529	1.2186	0.1977	0.0596	1.1738	0.1500	0.0452
120	1.2033	0.1761	0.0531	1.2760	0.2740	0.0826	1.1301	0.1829	0.0552
150	1.1817	0.1747	0.0527	1.1974	0.1990	0.0600	1.1803	0.1606	0.0484
180	1.1911	0.2054	0.0619	1.2083	0.1910	0.0576	1.1762	0.1639	0.0494
210	1.1560	0.1727	0.0521	1.2139	0.2032	0.0613	1.1318	0.1517	0.0458
240	1.2453	0.2783	0.0839	1.2402	0.1925	0.0580	1.1683	0.1559	0.0470

Table 11: Descriptive Statistics – APVEL – EOSRV

	Descriptive Statistics - APVEL EOSRV - CONDITION 3 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.3406	0.2840	0.0856	1.2073	0.1293	0.0390	1.1821	0.1618	0.0488
30	1.1462	0.1544	0.0465	1.1962	0.1957	0.0590	1.1788	0.1618	0.0488
60	1.1499	0.1618	0.0488	1.2092	0.1821	0.0549	1.1627	0.2107	0.0635
90	1.2063	0.2018	0.0608	1.1814	0.1960	0.0591	1.1836	0.1906	0.0575
120	1.1962	0.1809	0.0545	1.1937	0.1734	0.0523	1.2057	0.1824	0.0550
150	1.1813	0.1870	0.0564	1.1731	0.2094	0.0631	1.2210	0.2103	0.0634
180	1.1677	0.1764	0.0532	1.1818	0.1804	0.0544	1.2430	0.0981	0.0296
210	1.2426	0.2116	0.0638	1.1604	0.1951	0.0588	1.2779	0.2886	0.0870
240	1.1525	0.1175	0.0354	1.2170	0.1566	0.0472	1.2124	0.1568	0.0473

Table 12: Descriptive Statistics – APVEL – EOSRP

	Descriptive Statistics - APVEL EOSRP - CONDITION 4 (N=10)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.5012	0.3586	0.1134	1.5529	0.2265	0.0716	1.5956	0.3414	0.1080
30	1.4155	0.2463	0.0779	1.4699	0.3241	0.1025	1.4876	0.2685	0.0849
60	1.3603	0.1352	0.0428	1.4380	0.3397	0.1074	1.4494	0.2513	0.0795
90	1.3842	0.1978	0.0626	1.4091	0.2240	0.0708	1.4028	0.1809	0.0572
120	1.4502	0.3162	0.1000	1.4269	0.3396	0.1074	1.4190	0.2418	0.0765
150	1.3662	0.2605	0.0824	1.3848	0.3097	0.0979	1.5436	0.3383	0.1070
180	1.4285	0.2570	0.0813	1.3927	0.2915	0.0922	1.4697	0.2093	0.0662
210	1.4660	0.3245	0.1026	1.3627	0.3707	0.1172	1.4117	0.2113	0.0668
240	1.4644	0.2252	0.0712	1.4111	0.3505	0.1108	1.4575	0.2850	0.0901

Table 13: Descriptive Statistics – APVEL – ECSRP

	Descriptive Statistics - APVEL ECSRP - CONDITION 5 (N=9)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	3.0886	0.8067	0.2689	3.3470	0.6008	0.2003	2.8449	0.7057	0.2352
30	2.9687	0.7127	0.2376	2.8753	0.3765	0.1255	3.0016	0.4899	0.1633
60	2.7472	0.6129	0.2043	3.1191	0.4676	0.1559	2.8624	0.6391	0.2130
90	2.5937	0.5396	0.1799	2.7832	0.5715	0.1905	2.8113	0.5162	0.1721
120	2.7883	0.5950	0.1983	3.0829	0.2285	0.0762	2.7605	0.5155	0.1718
150	2.6918	0.4998	0.1666	2.7396	0.4179	0.1393	2.6897	0.3773	0.1258
180	2.7242	0.5753	0.1918	3.0681	0.8200	0.2733	2.8222	0.6620	0.2207
210	2.5814	0.4327	0.1442	2.6347	0.7749	0.2583	2.7865	0.5246	0.1749
240	2.5330	0.4045	0.1348	2.5686	0.5909	0.1970	2.8876	0.3290	0.1097

Table 14: Descriptive Statistics – APVEL – EOSRVP

	Descriptive Statistics - APVEL EOSRVP - CONDITION 6 (N=9)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	2.7400	0.7368	0.2456	2.8081	0.6559	0.2186	2.8836	0.4862	0.1621
30	2.5295	0.6544	0.2181	2.5057	0.4240	0.1413	2.6148	0.4122	0.1374
60	2.5383	0.5927	0.1976	2.6063	0.6078	0.2026	2.6411	0.3704	0.1235
90	2.7129	0.7976	0.2659	2.6020	0.7086	0.2362	2.4587	0.3230	0.1077
120	2.4087	0.4158	0.1386	2.3426	0.4302	0.1434	2.3731	0.2785	0.0928
150	2.4509	0.6386	0.2129	2.2546	0.4099	0.1366	2.6314	0.3575	0.1192
180	2.4634	0.5250	0.1750	2.4895	0.7024	0.2341	2.4380	0.3052	0.1017
210	2.4957	0.5899	0.1966	2.5073	0.4232	0.1411	2.4299	0.2578	0.0859
240	2.4879	0.3960	0.1320	2.2895	0.4100	0.1367	2.3105	0.3131	0.1044

Table 15: Descriptive Statistics – MLRMS – EO

	Descriptive Statistics - MLRMS EO - CONDITION 1 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.1550	0.0398	0.0115	0.1536	0.0602	0.0174	0.1279	0.0373	0.0108
30	0.1603	0.0447	0.0129	0.1453	0.0528	0.0152	0.1361	0.0430	0.0124
60	0.1725	0.0504	0.0145	0.1647	0.0689	0.0199	0.1965	0.1084	0.0313
90	0.1760	0.0495	0.0143	0.1720	0.0711	0.0205	0.1712	0.0798	0.0230
120	0.1778	0.0560	0.0162	0.1809	0.1165	0.0336	0.1433	0.0430	0.0124
150	0.1637	0.0435	0.0126	0.1618	0.0680	0.0196	0.1501	0.0441	0.0127
180	0.1648	0.0325	0.0094	0.1776	0.0661	0.0191	0.1505	0.0616	0.0178
210	0.1717	0.0510	0.0147	0.1545	0.0647	0.0187	0.1611	0.0660	0.0191
240	0.1868	0.0473	0.0137	0.1798	0.0565	0.0163	0.2004	0.1114	0.0322

Table 16: Descriptive Statistics – MLRMS – EC

	Descriptive Statistics - MLRMS EC - CONDITION 2 (N=10)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.1825	0.0454	0.0143	0.1374	0.0367	0.0116	0.1330	0.0366	0.0116
30	0.1845	0.0400	0.0126	0.1660	0.0551	0.0174	0.1485	0.0450	0.0142
60	0.2087	0.0399	0.0126	0.1850	0.0616	0.0195	0.1488	0.0377	0.0119
90	0.1912	0.0365	0.0115	0.1741	0.0540	0.0171	0.1554	0.0501	0.0158
120	0.2161	0.0663	0.0210	0.1692	0.0547	0.0173	0.1634	0.0486	0.0154
150	0.2037	0.0660	0.0209	0.1767	0.0548	0.0173	0.1674	0.0527	0.0167
180	0.2123	0.0594	0.0188	0.1759	0.0416	0.0132	0.2044	0.0687	0.0217
210	0.2124	0.0689	0.0218	0.1962	0.0757	0.0239	0.1867	0.0589	0.0186
240	0.2165	0.0665	0.0210	0.1915	0.0681	0.0215	0.1970	0.0503	0.0159

Table 17: Descriptive Statistics – MLRMS – EOSRV

	Descriptive Statistics - MLRMS EOSRV - CONDITION 3 (N=10)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.1972	0.0649	0.0205	0.1632	0.0745	0.0236	0.1372	0.0528	0.0167
30	0.2019	0.0860	0.0272	0.1961	0.1100	0.0348	0.1576	0.0403	0.0127
60	0.1837	0.0548	0.0173	0.1675	0.0549	0.0174	0.1779	0.0787	0.0249
90	0.1926	0.0478	0.0151	0.2285	0.1384	0.0438	0.1818	0.0500	0.0158
120	0.1963	0.0540	0.0171	0.2073	0.0500	0.0158	0.1441	0.0465	0.0147
150	0.1749	0.0405	0.0128	0.1781	0.0675	0.0213	0.1829	0.0453	0.0143
180	0.2084	0.0478	0.0151	0.2019	0.0591	0.0187	0.2460	0.1348	0.0426
210	0.2145	0.0907	0.0287	0.2096	0.0706	0.0223	0.2153	0.1038	0.0328
240	0.2214	0.0737	0.0233	0.2065	0.0555	0.0175	0.2167	0.0822	0.0260

Table 18: Descriptive Statistics – MLRMS – EOSRP

	Descriptive Statistics - MLRMS EOSRP - CONDITION 4 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.2526	0.0963	0.0290	0.1835	0.0549	0.0166	0.1725	0.0448	0.0135
30	0.2829	0.1138	0.0343	0.2006	0.0687	0.0207	0.1990	0.0545	0.0164
60	0.2426	0.1000	0.0302	0.1935	0.0529	0.0159	0.1915	0.0829	0.0250
90	0.2375	0.0585	0.0177	0.2342	0.1494	0.0450	0.2041	0.0598	0.0180
120	0.2695	0.0839	0.0253	0.2301	0.0791	0.0238	0.2261	0.0946	0.0285
150	0.2345	0.0868	0.0262	0.2024	0.0557	0.0168	0.2407	0.1010	0.0305
180	0.2831	0.0718	0.0216	0.2348	0.1007	0.0304	0.2249	0.0723	0.0218
210	0.2881	0.1189	0.0359	0.2493	0.1318	0.0397	0.2621	0.0936	0.0282
240	0.2710	0.0722	0.0218	0.2328	0.0888	0.0268	0.2650	0.1177	0.0355

Table 19: Descriptive Statistics – MLRMS – ECSRP

	Descriptive Statistics - MLRMS ECSRP - CONDITION 5 (N=10)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.3328	0.1032	0.0326	0.3034	0.0817	0.0258	0.2883	0.1067	0.0337
30	0.3650	0.1208	0.0382	0.3044	0.0951	0.0301	0.3053	0.0730	0.0231
60	0.3915	0.1081	0.0342	0.3734	0.1068	0.0338	0.3414	0.1082	0.0342
90	0.3573	0.0960	0.0303	0.3171	0.0895	0.0283	0.3798	0.1393	0.0441
120	0.3953	0.0947	0.0300	0.3575	0.1446	0.0457	0.3456	0.1309	0.0414
150	0.3455	0.0792	0.0251	0.3415	0.1097	0.0347	0.3232	0.1117	0.0353
180	0.3935	0.1216	0.0385	0.4062	0.1483	0.0469	0.3672	0.1433	0.0453
210	0.3400	0.0891	0.0282	0.3679	0.1071	0.0339	0.3503	0.1130	0.0357
240	0.3549	0.0997	0.0315	0.3439	0.0905	0.0286	0.3520	0.1024	0.0324

Table 20: Descriptive Statistics – MLRMS – EOSRVP

	Descriptive Statistics - MLRMS EOSRVP - CONDITION 6 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.2971	0.0715	0.0216	0.2702	0.0851	0.0257	0.2508	0.0973	0.0293
30	0.3178	0.1244	0.0375	0.2690	0.0629	0.0190	0.2857	0.0870	0.0262
60	0.3238	0.1069	0.0322	0.3070	0.0718	0.0217	0.2936	0.0924	0.0279
90	0.3373	0.1254	0.0378	0.3270	0.1226	0.0370	0.2866	0.1036	0.0312
120	0.3075	0.0799	0.0241	0.3149	0.0909	0.0274	0.3220	0.1127	0.0340
150	0.3244	0.1015	0.0306	0.3267	0.0785	0.0237	0.3344	0.0813	0.0245
180	0.3415	0.1182	0.0356	0.3571	0.1165	0.0351	0.3434	0.0923	0.0278
210	0.2954	0.0969	0.0292	0.3496	0.0778	0.0235	0.2929	0.0974	0.0294
240	0.3373	0.1105	0.0333	0.2951	0.0631	0.0190	0.3665	0.1616	0.0487

Table 21: Descriptive Statistics – MLVEL – EO

	Descriptive Statistics - MLVEL EO - CONDITION 1 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.7936	0.0746	0.0215	0.7593	0.0717	0.0207	0.7360	0.0599	0.0173
30	0.7518	0.0509	0.0147	0.7608	0.0997	0.0288	0.7346	0.0574	0.0166
60	0.7296	0.0470	0.0136	0.7765	0.1105	0.0319	0.7424	0.1094	0.0316
90	0.7427	0.0827	0.0239	0.7522	0.0855	0.0247	0.7543	0.0948	0.0274
120	0.7404	0.0520	0.0150	0.7496	0.0664	0.0192	0.7496	0.0826	0.0239
150	0.7260	0.0628	0.0181	0.7293	0.0643	0.0186	0.7159	0.0728	0.0210
180	0.7319	0.0773	0.0223	0.7514	0.0706	0.0204	0.7138	0.0402	0.0116
210	0.7637	0.0773	0.0223	0.7500	0.0822	0.0237	0.7146	0.0682	0.0197
240	0.7540	0.0592	0.0171	0.7282	0.0673	0.0194	0.7599	0.1164	0.0336

Table 22: Descriptive Statistics – MLVEL – EC

	Descriptive Statistics - MLVEL EC - CONDITION 2 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.8258	0.0847	0.0256	0.8304	0.0740	0.0223	0.8068	0.0700	0.0211
30	0.8269	0.0768	0.0232	0.7964	0.0896	0.0270	0.7820	0.0928	0.0280
60	0.8152	0.0663	0.0200	0.8063	0.0949	0.0286	0.7718	0.0700	0.0211
90	0.8028	0.0694	0.0209	0.8024	0.0692	0.0209	0.7778	0.0729	0.0220
120	0.8003	0.0719	0.0217	0.8048	0.0915	0.0276	0.7596	0.0620	0.0187
150	0.7939	0.0766	0.0231	0.7941	0.0881	0.0266	0.7901	0.0840	0.0253
180	0.8011	0.0832	0.0251	0.8180	0.1038	0.0313	0.7877	0.0875	0.0264
210	0.8106	0.0834	0.0251	0.8028	0.0875	0.0264	0.7655	0.0639	0.0193
240	0.8055	0.1046	0.0315	0.8323	0.0878	0.0265	0.7873	0.0638	0.0192

Table 23: Descriptive Statistics – MLVEL – EOSRV

	Descriptive Statistics - MLVEL EOSRV - CONDITION 3 (N=12)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.8316	0.0903	0.0261	0.7775	0.0594	0.0171	0.7785	0.0546	0.0158
30	0.7921	0.0786	0.0227	0.7985	0.1126	0.0325	0.7828	0.0672	0.0194
60	0.7825	0.0701	0.0202	0.7919	0.1075	0.0310	0.7587	0.0756	0.0218
90	0.7876	0.0732	0.0211	0.7801	0.0815	0.0235	0.7667	0.0851	0.0246
120	0.7943	0.0935	0.0270	0.7953	0.0854	0.0247	0.7662	0.0993	0.0287
150	0.8002	0.0903	0.0261	0.7872	0.0909	0.0262	0.7810	0.0996	0.0288
180	0.7852	0.0878	0.0253	0.8016	0.0946	0.0273	0.7911	0.0811	0.0234
210	0.8071	0.0796	0.0230	0.7920	0.0825	0.0238	0.7964	0.0940	0.0271
240	0.8048	0.0588	0.0170	0.8147	0.0967	0.0279	0.7799	0.0793	0.0229

Table 24: Descriptive Statistics – MLVEL – EOSRP

	Descriptive Statistics - MLVEL EOSRP - CONDITION 4 (N=8)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	0.8833	0.0531	0.0188	0.8868	0.0685	0.0242	0.8635	0.0413	0.0146
30	0.8611	0.0719	0.0254	0.8541	0.1096	0.0388	0.8422	0.0448	0.0158
60	0.8463	0.0595	0.0210	0.8531	0.0767	0.0271	0.8312	0.0468	0.0166
90	0.8655	0.0479	0.0169	0.8480	0.0851	0.0301	0.8452	0.0509	0.0180
120	0.8989	0.0766	0.0271	0.8701	0.0836	0.0296	0.8358	0.0695	0.0246
150	0.8633	0.0962	0.0340	0.8484	0.0886	0.0313	0.8499	0.0668	0.0236
180	0.8842	0.1266	0.0447	0.8485	0.0819	0.0290	0.8416	0.0342	0.0121
210	0.9002	0.0945	0.0334	0.8465	0.1038	0.0367	0.8491	0.0437	0.0155
240	0.8959	0.0784	0.0277	0.8564	0.0739	0.0261	0.8570	0.0593	0.0209

Table 25: Descriptive Statistics – MLVEL – ECSRP

	Descriptive Statistics - MLVEL ECSRP - CONDITION 5 (N=10)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.2056	0.1345	0.0425	1.2544	0.1753	0.0554	1.1621	0.2286	0.0723
30	1.2061	0.1562	0.0494	1.1577	0.2243	0.0709	1.1186	0.1017	0.0322
60	1.1590	0.1351	0.0427	1.2045	0.1165	0.0368	1.1017	0.1274	0.0403
90	1.1228	0.1092	0.0345	1.1145	0.0777	0.0246	1.1099	0.1360	0.0430
120	1.1809	0.1840	0.0582	1.1553	0.1481	0.0468	1.1175	0.1410	0.0446
150	1.1343	0.1513	0.0478	1.1014	0.1147	0.0363	1.0827	0.1191	0.0377
180	1.1611	0.2239	0.0708	1.1825	0.1930	0.0610	1.0691	0.1666	0.0527
210	1.1070	0.1261	0.0399	1.1485	0.2155	0.0682	1.0861	0.1437	0.0454
240	1.1313	0.1757	0.0556	1.1019	0.1867	0.0590	1.1141	0.1324	0.0419

Table 26: Descriptive Statistics – MLVEL – EOSRVP

	Descriptive Statistics - MLVEL EOSRVP - CONDITION 6 (N=11)								
	LOW			TACTICAL			WORK		
	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error	Mean	Std. Dev	Std. Error
PRE	1.0540	0.1491	0.0450	1.1593	0.2071	0.0624	1.0412	0.1299	0.0392
30	1.0462	0.1518	0.0458	1.0693	0.1251	0.0377	1.0302	0.0975	0.0294
60	1.0465	0.1325	0.0400	1.0976	0.1860	0.0561	1.0107	0.1343	0.0405
90	1.0600	0.1847	0.0557	1.0881	0.1483	0.0447	1.0344	0.1967	0.0593
120	1.0112	0.1329	0.0401	1.0260	0.1132	0.0341	1.0058	0.1187	0.0358
150	1.0402	0.1736	0.0523	1.0387	0.1298	0.0391	1.0410	0.1467	0.0442
180	1.0316	0.1608	0.0485	1.0886	0.1792	0.0540	1.0233	0.1183	0.0357
210	1.0365	0.1552	0.0468	1.0635	0.1712	0.0516	1.0179	0.1297	0.0391
240	1.0703	0.1229	0.0371	1.0263	0.1349	0.0407	1.0290	0.1150	0.0347

APPENDIX: B

RECRUITMENT SCRIPT

“RECRUITMENT SCRIPT”

(Verbal, in person: A brief version of the consent document.)

My name is Mr. Harish Chander, Dr. Garner, Waddell or Wade, a (*graduate student, faculty member, …*) from the Department of HESRM at the University of Mississippi. I would like to invite you to participate in my research study, the change of postural stability over extended periods of time from walking in different types of industry standard footwear.

You may participate if you do not have any musculoskeletal disorders or medical conditions that may be aggravated by exercise. Please do not participate if you have an injury to the lower or upper extremities, or balance disorders. As a participant, you will be asked to walk along a flat surface on for separate days. You will be asked to walk for a period of 4 hours, while you are tested every 30 minutes for balance and muscle fatigue changes. The total time will be 4.5 hours per sessions.

This exercise may cause muscle soreness, or possible falls due to lack of balance. However, injuries and falls are highly unlikely. Subjects are responsible for any and all medical costs that may result from injury during or related to the study. To complete the study you will be required to attend a total of 4 sessions each lasting 4.5hr each. Your decision whether or not to participate will not affect your course credit or university standing.

If you are interested in participating, please place your name and email address on the signup sheet that is being passed around. A member of the Biomechanics lab staff will be in contact to set up testing times. Do you have any questions now? If you have questions later, please contact me following class.

APPENDIX: C

INFORMED CONSENT

INFORMED CONSENT

Consent to Participate in an Experimental Study
Title: The effect of extended durations of walking on ballast rock and a sloped surface on postural stability

Investigator
John C. Garner, Ph.D., CSCS
Department of Health, Exercise Science, and Recreational Management
University of Mississippi – 215 Turner Center University, MS 38677
Tel: (662) 915-7573

Co-Investigator
Harish Chander, B.P.T
Department of Health, Exercise Science, and Recreational Management
University of Mississippi – 215 Turner Center University, MS 38677
Tel: (662) 915-7211

Description
You are being asked to participate in a research study for the purpose of investigating the effects of extended durations of walking/standing on a hard flat surface while wearing different types of footwear. The long-term goal of this proposed research is to minimize the risk of falling and injuries in individuals working in challenging work environments for long periods of time, leading towards implementing appropriate intervention strategies. The use of this knowledge might improve injury prevention among the working population. In this study, we will focus on balance and muscle fatigue. More specifically, we will look at the changes in leg muscle fatigue and balance due to walking on in different types of shoes for long periods of time.

Your participation is voluntary. If you decide to participate in the research study, four visits are required to complete the testing. Each visit will last approximately 4.5 hours. During each visit, you will receive participate in an experimental procedure related to balance and walking (explained below).

You will participate in a total of four protocols during the course of the visits and be exposed to four different footwear types (one each visit). The types of footwear are: barefoot, flat sole/slip resistant (restaurant type), work boots (industry standard), and tactical boots (military or law enforcement).

Protocol 1) NeuroCom Equitest Balance procedures

Protocol 2) Muscle fatigue testing

When you come in for the testing session, a member of the research team will explain the overall goal of this study and emphasize the importance of walking as naturally as possible throughout the experiment. Each of the following will be completed with each of the footwear types.

<u>Balance Testing:</u>

1) You will be fitted into the safety harness which will be attached to the balance measuring equipment.
2) You will be asked to stand with you arms crossed in front of your body while the floor beneath you and the screen in front of you moves at a very slow rate.
3) This procedure will be done one time for a total of 2.5 minutes.
4) At the completion of the balance measurements, you will be removed from the safety harness.
5) This process will take approximately 3 minutes.
6) You will be escorted to the walking surface.

<u>Muscle testing:</u>

1) You will be asked to sit in a chair with both feet placed flat on the floor.
2) EMG electrodes will be placed on your leg corresponding to a muscle
2) At a given instructions on the muscle being tested, you will be asked to contract that muscle. The muscle consist of the ankle and knee muscles.
3) After muscle testing you will be asked to walk onto the assigned surface.
4) You will be asked to begin walking at your own pace.

This complete cycle will occur 9 times (0, 30, 60, 90, 120, 150, 180, 210, 240min), for each of the 3 surfaces. Your total time will be 5 hours per session for a total of 15 hours over a few weeks time.

As part of the research study, we will record video of your movements while walking. These images will consist of your face, body, and body movements while walking. The confidential CD will be kept indefinitely. Your name will not be recorded in any way on the CD. Only your subject number and date of testing will be written on the CD's label. Unless you give separate permission below, only the investigators associated with this study will have access to the CD. The CD and any identifiable material will be stored in a locked filing cabinet within the Applied Biomechanics and Ergonomics Laboratory, in which only the investigators will have access. This recording will be studied by the research team for use in the research project. We would also like you to indicate below to what other uses of these digital recordings you are willing to consent. In each of the uses listed below, images, such as your face and movements recorded on the CD will be used for the purpose of describing the research procedures and in discussion of research findings. If you are not willing to consent to other uses of the digital recordings, you are still eligible to participate in the study. We will only use the CDs in the ways to which you agree. In any use of these CDs, your name would not be identified; however, such use does present a risk for loss of confidentiality because your image will not be altered to prevent your identification. These CDs will be kept indefinitely.

1. The digital recordings can be shown to subjects in other experiments…______
Initials
2. The digital recordings can be used for scientific publications………… ______
Initials
3. The digital recordings can be shown at meetings of scientists………… ______
Initials
4. The digital recordings can be shown in classrooms to students………. ______
Initials
5. The digital recordings can be shown in public presentations to nonscientific groups…………………………………………………….______
Initials
6. The digital recordings can be used on television……………………… ______
Initials
7. The digital recordings can be used on a public website maintained by the research group………………………………………………..… ______
Initials

Risks and Benefits

Adverse events of this research study are listed with the following risk categories:

"Rare" – occur in less than 1% or less than 1 in 100 people

Balance and Gait: Injury risk is rare. The only possible risk would be due to an unexpected fall. The incidence of falling is rare and will be protected against through the presence of laboratory staff. There are other potential risks for injuries even if the subject does not fall onto the ground: Muscle pull, muscle tear, skin abrasion, chafing, and sudden movement-related injuries (e.g. being jerked), which may occur in the event of an equilibrium loss. At all times during balance testing, the subject will wear a safety harness designed to eliminate the risk of falling during the balance testing protocol.

You will likely receive no direct benefit from taking part in this research study. Should the testing procedures performed yield results that are abnormal, e.g. abnormal balance, abnormal walking, you will be advised. If you decide to speak to your physician, it will be your responsibility set up an appointment with him/her. The results will be available at no cost, should you or your physician request them.

Confidentiality
Any information about you obtained from or for this research study will be kept as confidential (private) as possible. The records identifying your name will be (1) stored in a locked file cabinet and/or in a password-protected computer file, (2) kept separate from the rest of the research records, and (3) be accessible to only the researchers listed on the first page of this form and their staff. Your identity on the other research records will be indicated by a case number rather than by your name. You will not be identified by name in any publication of the research results unless you sign a separate form giving your permission (release).

Right to Withdraw
Your participation in this research study, to include the use and disclosure of your identifiable information for the purposes described above, is completely voluntary. (Note, however, that if you do not provide your consent for the use and disclosure of your identifiable information for the purposes for the use of the recordings described above, you will still be allowed to participate in the research study, and the recordings will not be used for anything other than analysis by the staff.) Whether or not you provide your consent for participation in this research study will have no affect on your current or future relationship with the University of Mississippi.

You may withdraw, at any time for any reason, your consent for participation in this research study, to include the use and disclosure of your identifiable information for the purposes described above. This voluntary withdraw can be for any reasons such as: physical discomfort of any kind, emotional distress, feeling uneasy about the testing procedure, time constraints, and/or lack of interests in participation. Any reason for which you feel as though you do not wish to continue can be a means of discontinuing the study. Any and all identifiable research information (CDs) recorded for, or resulting from, your participation in this research study prior to the date that you formally withdrew your consent will be destroyed immediately.

If you start the study and decide that you do not want to finish, all you have to do is to tell Dr. John Garner, Dr. Chip Wade, Dr. Dwight Waddell, or Mr. Harish Chander in person, by letter, or by telephone at the Department of Health, Exercise Science, and Recreational Management, 215 Turner Center, The University of Mississippi, University MS 38677, or 915-5561. Whether or not you choose to participate or to withdraw will not affect your standing with the Department of Health, Exercise Science, and Recreational Management, or with the University, and it will not cause you to lose any benefits to which you are entitled.

IRB Approval
This study has been reviewed by The University of Mississippi's Institutional Review Board (IRB). The IRB has determined that this study fulfills the human research subject protections obligations required by state and federal law and University policies. If you have any questions, concerns or reports regarding your rights as a participant of research, please contact the IRB at (662) 915-7482.

Statement of Consent
I have read the above information. I have been given a copy of this form. I have had an opportunity to ask questions, and I have received answers. I consent to participate in the study.

____

Signature of Participant Date

____

Signature of Investigator Date

Statement of Consent to be Contacted for Future Studies
The staff of the Applied Biomechanics and Ergonomics Laboratory and/or Body Composition Laboratory may be of interested in contacting you to participate in future studies. Signing below allows us to contact you to contact you with information on future studies.

____

Signature of Participant Date

____

Signature of Investigator Date

NOTE TO PARTICIPANTS: DO <u>NOT</u> SIGN THIS FORM

IF THE IRB APPROVAL STAMP ON THE FIRST PAGE HAS EXPIRED.

APPENDIX: D

RECRUITMENT FLYER

Men
18 to 45 years old

The Applied Biomechanics and Ergonomics Laboratory (ABEL) at the University of Mississippi is conducting a research study on how individuals walk in different types of footwear.

- Participants will be asked to come in for 4 visits.
- Sessions will last about 4.5 hours.

Must not have a heart condition
Must not have balance/dizziness complaints
Must not have joint movement problems
Must not have epilepsy, tremors, or a stroke
Must walk normally

If interested, call **(662) 915-7211**.
Ask for Mr. Harish Chander (UMABEL@olemiss.edu)
215 Turner Center, University of Mississippi

APPLIED BIOMECHANICS & ERGONOMICS LABORATORY (ABEL)

APPENDIX: E

OCCUPATIONAL FOOTWEAR TYPES

WORK BOOT: ANSI – Z41 – 1991 STANDARDS

TACTICAL BOOT

LOW-TOP SLIP-RESISTANT SHOE

VITA

Harish Chander, B.P.T, CMT,

Academic Records:

Bachelor of Physical Therapy The TN Dr.MGR Medical University, Chennai, India
Sree Balaji College of Physiotherapy
Work Concentration: Neuro-Developmental Therapy and Sports Injury Rehabilitation
Under-Grad Dissertation: The Effects of Cryotherapy and Massage in Delayed Onset of Muscle Soreness

Employment History:

2007 – 2008

Peadiatric Physical Therapist
Vinayaga Physio Point, Chennai, India
Appointments:

- Pediatric Neuro-Development Therapy
- Pain Relief of Musculoskeletal Injuries
- Rehabilitation of Orthopedic and Neurological Conditions

2008 – 2009

Fitness coordinator and Physical Therapist
Talwalkars Better Value Fitness Pvt. Ltd, Chennai, India
Appointments:

- Exercise Prescription and Testing
- Fitness Trainer
- Administrative official for promoting health & fitness

2009 – 2010

Student worker - Law Library, University of Mississippi, University, MS
Appointments:

- Itemization of book collections
- Inventory Maintenance
- Journal and Periodicals’ Maintenance

2010 – present Graduate Teaching and Research Assistant
Applied Biomechanics Laboratory, University of Mississippi, University, MS
Appointments:

- Gait, Posture and Footwear Biomechanics
- Ergonomics
- Undergraduate Biomechanics Laboratory Instructor
- Undergraduate Kinesiology Instructor
- Undergraduate Personal & Community Health Instructor