Evaluation of Circumferential Ankle Pressure as an Ergonomic Intervention to Maintain Balance Perturbed by Localized Muscular Fatigue of the Ankle Joint

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ABSTRACT

Application of pressure in the form of taping and bracing has been shown to improve proprioception, and inducing localized muscle fatigue at various musculatures has been shown to adversely affect postural control. However, the potential for pressure application to mitigate the effects of localized muscle fatigue on postural control has not yet been determined. This study investigated specifically the effects of circumferential ankle pressure (CAP) and induced ankle fatigue on postural control. Fourteen young participants (seven males and seven females) performed fatiguing sub-maximal isotonic plantar flexion exercises on an isokinetic dynamometer, in the absence and presence of a pressure cuff (60 mm Hg) used to apply CAP. Proprioceptive acuity (PA) was determined using a passive-active joint position sense test, with categorical scores (low or high PA) used as a covariate. Postural sway during quiet standing was assessed using a force platform both pre- and post-fatigue as well as in the absence and presence of CAP. Application of CAP resulted in larger postural sway in individuals with low PA, and reduced postural sway in individuals with high PA. Fatigue effects on postural sway in individuals with low PA were more substantial as compared to individuals with high PA. CAP was found to be ineffective in mitigating the effects of fatigue on postural sway in individuals with lower PA. As a whole, the results suggest a potential for CAP as an ergonomic intervention in controlling fatigue-related fall incidents, though conclusive recommendations for use are not justified.

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LIST OF ABBREVIATIONS

Abbreviation	Definition
AAE	Absolute angular error
AP	Antero-posterior direction
BOS	Base of Support
CAP	Circumferential Ankle Pressure
CNS	Central Nervous System
COM	Center of Mass
DFA	Detrended Fluctuation Analysis
DFAEX	Scaling exponent (α) for DFA
EA	Ellipse area
ERR	COP - COM or the "Error" Signal
FFT	Fast Fourier Transform
HRARA	Hurst rescaled adjusted range analysis
JC	Joint center
JMS	Joint motion sense
JPS	Joint position sense
LI	Leisure Index in the Habitual physical activity questionnaire
LMF	Localized muscle fatigue
MCOP	Mean COP position
MDPF	Median power frequency
MinMTtb	Minimum Modified Time-to-boundary
ML	Medio-lateral direction
MMTtb	Mean Modified Time-to-boundary
MPF	Mean power frequency
MTtb	Modified Time-to-boundary
MVC	Maximum voluntary contraction
MVELO	Mean velocity
PA	Proprioceptive acuity
PC/PCS	Postural control/Postural control system
PSR	Postural stability rating
PtPERR	Peak-to-peak of COP-COM
PVELO	Peak velocity
R	Resultant direction
RDIST	RMS distance
RMS	Root mean square
RMSAnTh	RMS of ankle angle
RMSERR	RMS of COP-COM
RMSHiTh	RMS of hip angle
SA	Sway area
SB	Stability boundary
SDA	Stability diffusion analysis
SI	Sports index in the Habitual physical activity questionnaire
STDEVTtb	Standard deviation of Time-to-boundary
STIFF	Stiffness
Th	Theta; covariance of hip and ankle angle
Ttb	Time-to-boundary
WI	Work index in the Habitual physical activity questionnaire

INTRODUCTION

Occupational falls are a major cause of contemporary concern. In 2003, falls accounted for 272,988 nonfatal injuries and this number rose to 300,000 by the end of 2004 (Department of Labor, 2005). Falls resulted in 691 occupational fatalities, of which 52.2 % (361) occurred in the construction industry (Department of Labor, 2005). In 2002, falls from heights accounted for 8% of total occupational falls, resulting in the second largest absence rate, a total of 14 days away from work (BLS, 2004). In 1999 alone, direct costs associated with workplace injuries resulting from falls were estimated to be $3.7 billion (Department of Labor, 2005). These statistics indicate that falls from heights, especially in the construction industry, not only continue to impede productivity and growth of revenue but also are a cause of substantial personal costs.

Slips, trips and loss of balance are primary causes of occupational falls (Hsiao & Simeonov, 2001). Collectively, these are considered as 'loss of balance' incidents. The review conducted by Hsiao and Simeonov (2001) highlighted various factors that contribute to occupational falls, which can be broadly categorized as environmental (visual and physical interactions), task related (load handling and fatigue) and personal. The review also identified that existing measures, such as implementation of safety codes, regulations and guidelines and the use of safety equipment (guardrail systems, covers, etc.), which are aimed primarily at *protection* have not been completely effective due to negligence or improper use. Thus there is a need for identification and control of factors contributing to fall risk, with increased focus on fall *prevention* research. Localized muscle fatigue is one such task related factor associated with fall risk. The review provides motivation for future studies to investigate the effects of localized muscle fatigue on loss of balance incidents and incorporate controls and interventions to prevent falls.

According to the Newton's First Law, an object is in a state of equilibrium (maintenance of balance) if the resultant forces and moments acting on it are zero. Although quiet standing seems like a fairly simple task, maintaining balance during standing involves complicated neuromuscular control processes. During quiet standing, the human body can be modeled as a multilink inverted pendulum with the upper part of the body comprising two-thirds of total body weight (Winter, 1995) while simultaneously receiving destabilizing forces from gravity and the external environment (Alexander, 1994; Maki & McIlroy, 1996; Yaggie & McGregor, 2002). Hence, maintaining an upright posture involves controlling an inherently unstable system, and this task is performed by the postural control system (PCS). PC is a continuous process of maintaining the body center of mass (COM) over its base of support (BOS), where BOS is the area of the body in contact with the ground (Alexander, 1994; Winter, 1995; Maki & McIlroy, 1996; Yaggie & McGregor, 2002).

PC (Figure 1) involves continuous, nonlinear and complex sets of activities performed by the central nervous system (CNS) to process inputs received from the visual (planning and organization of standing and/or locomotion), vestibular (sensing linear and angular accelerations) and somatosensory (sensing position and velocity of body segments) systems (Patla, 1997; Maki & McIlroy, 1996; Winter, 1995; Blaszczyk et al., 2003). The somatosensory system receives information on length and tension from diverse sensors located in the skin, tendons, muscle spindles, and Golgi tendon organs (Gribble & Hertel, 2004). In addition, the PC system receives anticipatory and compensatory feedforward commands from the CNS, which constantly updates the sensory information, in order to maintain an upright posture. Due to these feedback and feedforward commands to and from the CNS, non-linear sets of activities, and latencies involved, a constant upright posture is not possible to achieve. Instead, our body exhibits sway to varying extents, due to the continuous adjustments and compensations. PC is impaired if any one of the three inputs to the CNS is perturbed. In such a situation, postural sway is usually observed to increase (Maki & McIlroy, 1996; Allum et al., 2004).

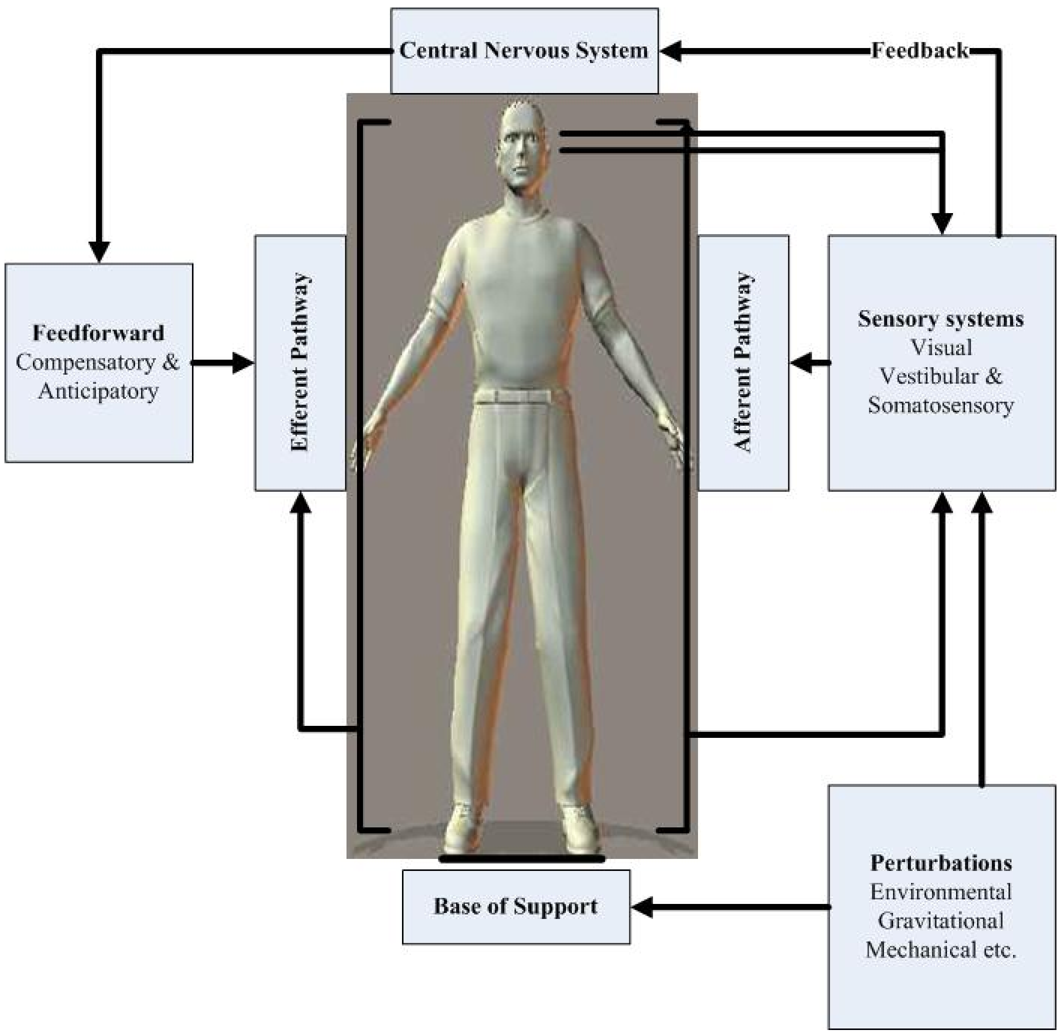

Figure 1. Schematic of processes involved in postural control

Position and movement (and forces causing these motions) of the whole body influence postural sway, hence, in order to establish sway as a valid indicator of balance and stability, most researchers have assessed sway during quiet standing. Previous studies have shown that individuals with a history of falls have a tendency to exhibit larger postural sway than non-fallers (Lichtenstein et al., 1988; Maki et al., 1994). Studies in the past have also shown that larger postural sway can be used as a predictor of risk of falls (Fernie et al., 1982). Postural sway can be assessed during quiet stance with or without subjecting one or more of the inputs to a perturbation (Loughlin et al., 2003).

During upright stance, the center of vertical reaction force, the center of pressure (COP), acts directly under the feet and reflects forces and torques required to keep the COM over the BOS (Loughlin et al., 2003). COP movement can be measured by having individuals stand on a force plate and plotting the COP path. This COP trajectory measured over a predetermined time interval, called a stabilogram, can be used to derive COP displacements in the antero-posterior (AP) and the medio-lateral (ML) directions (Alexander, 1994; Maki & McIlroy, 1996; Prieto et al., 1996; Blaszczyk et al., 2003; Loughlin et al., 2003). Many researchers in the past have used stabilograms to derive different statistical (Prieto et al., 1996; Loughlin et al., 2003), stochastic (Collins & De Luca, 1996), and kinetic (Riccio et al., 1992; Wegen et al., 2002) COP-based measures to better investigate the non-linear and complex behaviors involved in postural sway and control with or without a perturbation.

One underlying assumption supporting the use of COP based sway measures is that COP displays a time-invariant or stationary behavior. A stationary time series has a consistent mean and variance at least for the duration of the trial (Cao et al., 2004). However, this assumption has recently been challenged by various investigators, who have stated that COP is essentially a time-variant signal (Schumann et al., 1995; Ferdjallah et al., 1999; Loughlin & Redfern, 2001). If COP is a non-stationary signal, incorporating COP based measures that assume stationarity could lead to errors in analysis. Thus, one goal in the current study is to formally test the COP signal for stationarity.

Prior to conducting any analysis incorporating COP based sway measures, one must also consider the fact that these measures are highly correlated among each other. Prieto et al., (1996) showed that only a subset of these measures should be sufficient to analyze the performance and efficiency of the PCS. However, it must be noted here that very few studies have investigated which measures might be better suited to investigate the effects of a particular perturbation and its affects on the PCS. Due to the lack of evidence on the choice of measures, and specifically with respect to fatigue as a perturbation, this study incorporated a wide variety of sway measures. A correlation analysis was performed to investigate which measures might be highly correlated with each other and at the same time are effective in identifying the effects of fatigue on postural sway.

Localized muscle fatigue (LMF), defined as a reduction in force generating capacity (Corbeil et al., 2003), impairs proprioceptive (position perception) inputs, by adversely affecting muscle spindle activity and disrupting mechanoreceptor (somatosensory) feedback (Johnston et al., 1996; Voight et al., 1996). Previous studies have shown that LMF of the ankle increases postural sway in the AP and ML directions (Lundin et al., 1993; Vuillerme et al., 2002; Yaggie

& McGregor, 2002; Gribble & Hertel, 2004). For example, one study (Yaggie & McGregor, 2002) found that during quiet unilateral stance, total sway and sway in the ML direction increased after fatigue was induced isokinetically at the ankle. Another study (Gribble & Hertel, 2004) reported that, during quiet single leg stance, postural control (assessed using COP velocity) was adversely affected in the AP direction but not in the ML direction after fatigue was induced at the ankle. On the contrary, Adlerton and Moritz (1996), in a study investigating the effect of calf muscle fatigue on standing balance, reported that the fatiguing exercises did not increase postural sway. In this latter study, however, fatigue was induced by having individuals perform calf-raises on tip-toe, indicating a lack of control with regards to force and range of motion. From the reports summarized here, it could be assumed that fatigue deleteriously affects postural sway. Impairment of the somatosensory system by injury or a perturbation such as localized muscle fatigue (LMF) might perturb the sensory inputs to the CNS (Figure 1). These impairments, in turn, might lead to incorrect predictive feedforward and reactive feedback commands, through the efferent pathways to the motor mechanisms involved in postural control. In addition to this, fatigue also affects the efferent pathway (by affecting motor performance) of the postural control system directly (Enoka & Stuart, 1992). Thus, from the review of concepts highlighted here, it is evident that localized muscle fatigue perturbs muscle spindle structure, motor unit firing and the proprioceptive system, thereby likely compromising the PCS.

Proprioception is the sum of joint position sense (JPS) and kinesthesia. JPS or joint proprioception is defined as the positional sense of one's own joints in the absence of vision (Skinner et al., 1984; Voight et al., 1996; Tsang & Hui-Chan, 2003). Joint proprioception is commonly assessed quantitatively in terms of joint repositioning error. Joint repositioning error

(Voight et al., 1996; Batavia et al., 1999; Halseth et al., 2000; Miura et al., 2004; You et al., 2004) can be obtained by having individuals move their joints in order to reproduce a certain angular position which is either predefined or randomly selected. Multiple assessments are collected and absolute angular error (AAE) is calculated by taking the difference between the preset position and the reproduced position. Based on the AAE, individuals can be categorized as having low or high proprioceptive acuity (PA). Kinesthesia or joint motion sense (JMS) is assessed by detecting the threshold to passive movement, and tests the ability to detect relatively slow passive joint movement. Both JPS and JMS are methods of assessing the afferent pathways. Skinner et al., (1986) reported that LMF induced at the knee joint significantly affected JPS but not JMS, whereas Carpenter et al., (1998) found that shoulder fatigue significantly affected JMS. Thus, any compromise to proprioception, such as following fatigue, can be expected to increase postural sway and deteriorate balance.

Muscle, joint and cutaneous receptors constitute the proprioceptive system. Previously it was assumed that only joint and cutaneous receptors provide information about position and velocity. This theory was replaced by a more recent focus on muscle spindles and muscle receptors. The role of muscle receptors in joint positioning and motion was revisited, and researchers argued that muscle spindle receptors were primarily associated with position and velocity information, however joint and cutaneous receptors did play a role. Joint receptors were considered to fire only at the extreme movements or towards the beginning and end ranges of motion. It has also been argued that the firing of joint receptors, especially the Ruffini endings, Pacinian corpuscles and Golgi tendon organs, is dependent on whether the movement is active or passive (Schmidt & Lee, 1993; Lephart et al., 1998). Cutaneous receptors are more sensitive to pressure and pain sensations. Muscle spindle receptors are sensory gamma motoneurons present in the equatorial region (muscle belly) of the muscle and are sensitive to the change in position and velocity and these motoneurons are innervated by the onset of stimulus, thereby affecting proprioception (Schmidt & Lee, 1999).

Taping, bracing, applying pressure to or stretching the skin can stimulate the mechanoreceptors present in the muscle spindles (Simoneau et al., 1997; Batavia et al., 1999; Halseth et al., 2004). These studies report contradictory findings on the benefits of taping or bracing the ankle joint. For example, studies investigating the effects of taping or bracing the ankle joint (Robbins et al., 1995; Heit et al., 1996; Simoneau et al., 1997) reported that applying pressure at the ankle joint improved proprioception. One study (Simoneau et al., 1997) reported that taping improved joint proprioception while sitting but not while standing. Another study (Halseth et al., 2004) investigating the effects of specific taping procedure (Kinesio taping) on the ankle joint found no differences between tape and no-tape conditions on proprioception.

Circumferential pressure around joints is a relatively new technique involving application of pressure, which in turn provides stimulation, to the cutaneous receptors, joint mechanoreceptors and muscle spindle proprioceptors around that joint (Batavia et al., 1999). A preliminary investigation on the effects of circumferential ankle pressure (CAP) on PA, active stiffness, and postural stability (You et al., 2004) revealed that CAP improved PA, active stiffness, and postural stability in individuals with low PA. You et al. (2004) also reported that CAP did not provide external mechanical support to the ankle joint, thereby arguing that all the changes observed by application of CAP were due to neuromuscular factors. Proprioceptive acuity was assessed as the absolute joint position sense error obtained from an active-active joint sense test protocol. Due to a lack of evidence on the effects of CAP on postural stability, and whether it can be an effective intervention in situations where the ankle joint has been perturbed by fatigue, further research is warranted in this area. The current study specifically investigated whether CAP can help maintain postural stability after it has been perturbed by fatigue, using sway as an indirect assessment of stability.

RESEARCH OBJECTIVES

Rationale

As previously mentioned, falls from heights are prevalent in industry, and LMF may be an important factor contributing towards falls. Fall prevention research should be aimed at preventing rather than protecting against falls as has been suggested in a recent review (Hsiao & Simenov, 2001). LMF at the ankle can be expected to cause impairment of both the afferent and efferent pathways involved, thereby impairing PC (Gribble & Hertel, 2004). Interventions recommending application of taping and/or bracing to apply pressure on mechanoreceptors of the skin have provided conflicting results. Recent studies have shown that applying circumferential joint pressure might be better than applying taping or braces to stimulate mechanoreceptors, due to better control of pressure application (Batavia et al., 1999; You et al., 2004). There are a limited number of studies investigating the effects of CAP on postural stability and whether it could be an effective intervention to a balance perturbation.

Previous studies have shown larger postural sway (measured as area of the stabilogram, and distance from the mean COP) among fallers (individuals with a history of or tendency to fall) as compared to non-fallers (Lichtenstein et al., 1988; Maki & McIlroy, 1996). Additionally there is anecdotal evidence suggesting the role of proprioception for proper functioning of the PCS (Lord et al., 1993). In the current study, it is hypothesized that CAP will be effective in reducing postural sway in individuals belonging to the low PA group, thereby helping these individuals maintain balance.

The rationale for the current study was to investigate whether CAP can be used as an effective ergonomic intervention to help maintain balance that has been perturbed by fatigue. For application of CAP to be an effective preventive strategy in controlling fatigue-related falls, it has to be used during the work activities. Thus, it would be advisable for workers (e.g. roofers) to apply CAP during daily work activities. In light of this argument, the exercise protocol required participants to wear CAP during the entire experiment, including the exercise. From a pilot study conducted previously, it was found that the endurance time (duration of time the exercise was conducted) for conditions with and without CAP was similar. In the current study, sway trials were collected during static standing with eyes-closed in a closed and quiet room, thereby eliminating visual and auditory feedbacks to the CNS. Vestibular input to the CNS cannot be impaired or affected by the proposed experimental procedure. Since visual input has been eliminated and vestibular input is unaffected, the current study aimed primarily at investigating (1) whether localized neuromuscular fatigue impairs the somatosensory input and/or the efferent motor mechanisms, thereby adversely affecting balance (as measured using stabilograms), and (2) whether stimulation to the somatosensory system, via CAP can help maintain balance.

PRIMARY HYPOTHESES

1. CAP will be more effective in reducing postural sway (quantified using COP-based sway measures) in individuals with low PA during upright stance.

   Null hypothesis H_0: No interactive effects of CAP and PA

2. Fatigue affects on postural sway in individuals with low PA will be more substantial as compared to other individuals during a upright stance.

   Null hypothesis H_0: No interactive effect of fatigue and PA

3. CAP will be more effective in maintaining postural stability after it has been perturbed by the effect of localized muscle fatigue in individuals with low PA.

   Null hypothesis H_0: No interactive effect of CAP, fatigue and PA

RESEARCH DESIGN AND METHODS

The current study is an extension of an ongoing research project addressing the effects of localized muscle fatigue on postural control when fatigue is induced at different joints, namely: at the ankle, knee, lower back and shoulder. The current study investigated the effectiveness of CAP as an ergonomic intervention in reducing the effects of localized fatigue, induced at the ankle joint, on postural control.

PARTICIPANTS

Due to the lack of evidence in existing literature addressing the issue of combined effects of CAP or taping, localized muscle fatigue and PA on postural sway, a formal power analysis could not be performed on the hypotheses being tested in the current study. Instead, a power approach based on existing data obtained from fallers and non-fallers was chosen to determine an adequate number of participants. As already mentioned, fallers tend to exhibit greater postural sway than non-fallers. For detecting the main effects of CAP with and without fatigue, values of a COP based sway measure (RMS distance in ML direction) for fallers (3.4 ± 2.3 mm) and non-fallers (2.1 ± 0.9 mm) were obtained from an earlier study (Maki & McIlroy, 1996). Weighted averages for mean and standard deviation were taken as the data was obtained from unequal sample sizes. The non-centrality parameter (δ) for a one factor, two level study is given as:

$$\delta = \frac{|\mu_1 - \mu_2|}{\sigma\sqrt{\frac{2}{n}}} \tag{1}$$

where, σ is the standard deviation of error terms, μ_1 and μ_2 are means of two populations of fallers and non-fallers and n is the sample size.

Table 1 presents power vs. n derived from this data.

Table 1. Power approach for sample size estimation

n	df	Δ	Power (1 – β)
2	1	1.655172	0.07
4	3	2.340767	0.29
6	5	2.866843	0.67
8	7	3.310345	0.73
10	9	3.701078	0.94
12	11	4.054328	0.95
14	**13**	**4.379175**	**0.96**
16	15	4.681535	1.00

Note: $\mu_1 - \mu_2 = 2.877$, $\sigma = 1.737$ (Maki & McIlroy, 1996) and Power for $\alpha = 0.05$

The current study is an exploratory investigation on the combined effects of CAP, localized muscle fatigue and PA on postural sway. Although, the sample size using the approach above might be too small to detect any interesting effects outlined in the hypotheses, the study aims at highlighting the importance of CAP in maintaining balance and attempts to provide a preliminary basis for future research investigations in the area of joint activation via applying pressure, and indicating how it might affect postural sway differently in various individuals. Simultaneously, the study also aims at providing a basis for treating pressure application as an intervention towards mitigating the effects of applied perturbations (especially localized muscle fatigue) and eventually achieving better postural control (and thereby indirectly providing better balance). The sample size chosen was n = 14. It is assumed that this sample size will be adequate across all dependent measures being used in the study.

Since this is an exploratory study, it is assumed that employing a relaxed alpha level is justified in order to increase power for detecting interactive effects, which are the main focus as outlined in the hypotheses. Thus, $\alpha = 0.1$ and n = 14, is assumed adequate for detecting any important effects associated with these hypotheses.

Fourteen participants (7 males and 7 females), with no self-reported injuries, illnesses, musculoskeletal disorders or occurrences of falls in the past year, were selected from the local community. Other relevant demographics, such as age, dominant leg, gender, etc., were also collected using the form provided in Appendix A. Level of physical activity was assessed using a questionnaire which incorporates physical activity during work, sports, and leisure time (Baecke et al., 1982). The questionnaire uses a semi-continuous scale ranging from low (1) to high (2), and has three different subsections for work (work index WI), sports (sports index, SI) and leisure (leisure index, LI). These subsections can be registered as three distinct indices to assess physical activity level. The physical activity questionnaire has been shown to have good reliability and validity for both male and female participants (Folsom, et al., 1997). Table 2 presents participant demographics and physical activity scores.

Table 2. Participant characteristics

				Physical Activity Scores		
	Age	**Mass (kg)**	**Stature (cm)**	**WI**	**SI**	**LI**
Males	20.2 ± 2.3	77.2 ± 8.1	175.8 ± 5.5	2.0 ± 0.6	3.0 ± 0.2	3.1 ± 0.9
Females	19.9 ± 1.9	59.4 ± 8.2	165.8 ± 4.9	2.4 ± 0.4	2.8 ± 0.6	3.6 ± 0.6

EXPERIMENTAL PROCEDURES

Upon arrival, participants were informed about the experimental procedures and completed an informed consent procedure approved by the VT IRB. Participants were provided several practice sway trials to help familiarize them with standing on a force plate (AMTI OR6-7-1000, Watertown, Massachusetts, USA). The participants were then calibrated with respect to a Postural Stability Ratings (PSR) scale, which was used in the experiment as a subjective measure. For the upper extreme of the scale (100, completely stable), the participants stood with their eyes open, head facing straight ahead, feet shoulder width apart and holding on to a solid object. For the lower extreme (0, not at all stable), the participants stood with their eyes closed in an unilateral stance. During the sway trials, participants stood on the force plate for 75 seconds, maintaining an upright posture with their feet together, arms by their sides, head straight and eyes closed, and were instructed to "concentrate on standing as still as possible". The participants were also requested to “think about the PSRs during the period of the quiet stance”.

The trials were conducted in a closed room to eliminate any noise disturbance, and to reduce or eliminate the effects of changing visual and auditory inputs to the CNS. A sheet of paper was placed on top of the force plate for each participant and tape was placed parallel to the AP edges and 15 cm behind the center of the force plate. Participants aligned their heels with the tape while standing, and an outline of the feet (or BOS) was drawn to standardize the location of stance across trials as shown in the Figure 2 (Maki, 1994; Wegen et al., 2002).

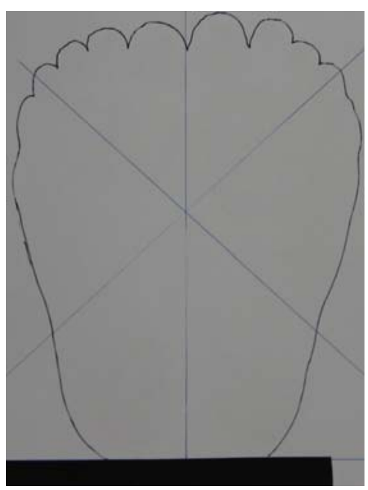

Figure 2. Foot placement and markings used to standardize stance during sway trials

Participants were requested to sit while resting between trials. Standing up on the force plate immediately after rest might induce changes in blood flow, thereby impacting postural sway patterns. Thus, participants were required to stand on the force plate 15 s prior to the onset of the trial to eliminate these transitional changes. Three sway baseline (pre-fatigue) trials were collected, with one minute in between. A total of 19 markers were placed bilaterally over the temple, chin, acromion, sternum, elbow joint centers (jc), wrist jc, iliac crest, knee jc, lateral malleoli, and the 5th metatarsal, and used, as described below, to estimate whole-body center of mass location.

A dynamometer (Biodex 3 Pro, Biodex Medical Systems Inc., Shirley, New York, USA) was used to induce fatigue in the ankle plantar flexors of the dominant foot. In situations where participants were not sure, the dominant foot was determined by asking participants which foot they used to kick a ball (Appendix A). Participants were then required to do calf-raises on the floor as warm-up exercise, consisting of 2 sets of 10 repetitions each.

Prior to the fatiguing exercises, maximum voluntary contractions (MVC) and sham exercises were conducted. MVC was conducted iso-kinetically for a minimum of five times at a speed of 60°/sec with an interval of 1 min in between for rest. Participants performed the exertions from approximately 15° dorsi-flexion to 30° plantar-flexion, as this is the range of motion most individuals can comfortably achieve (NASA, 1978). Participants were instructed to perform the exertions "as hard and as fast as they can" and were given non-threatening encouragement. MVC was recorded as the peak torque after adjusting for body segment and fixture masses. The dynamometer was brought back to the start range of motion passively, as the ankle attachment has a considerable mass and can cause fatigue in the dorsi-flexors. After this, participants were provided a rest period of 10 min.

Fatigue was induced using the dynamometer, with sub-maximal isotonic plantar-flexion exertions conducted at 60% of MVC, at a speed of 12 repetitions/min, through the same range of motion. Participants were instructed to start the exertions at the sound of an audio tone, and to try to reach the end of the range of motion at the sound of a second tone, in order to ensure that they maintained a consistent speed. Trials were terminated when the participant could not perform exertions for the entire range of motion in three consecutive trials. A final MVC trial was then conducted to determine the extent of fatigue. Participants were then asked to step on the force plate after 45 s and a sway trial was conducted. Due to the limitations of the experimental setup this was the shortest time for participants to step on the force plate after conducting exercises on the Biodex dynamometer.

In the CAP condition, the participants repeated the same protocol described earlier after applying 60 mm Hg pressure using a pediatric aneroid sphygmomanometer (Signature, Mabis Healthcare Inc., Lake Forest, Illinois, USA). The distal end of the cuff was aligned with the proximal end of the medial and lateral malleoli (You et al., 2004) of the dominant foot (Figure 3).

Figure 3. CAP is applied using an aneroid sphygmomanometer

DATA COLLECTION, PROCESSING & REDUCTION

Each sway trial lasted for 75 s and the first 10 s and the last 5 s were removed, providing a sample duration of 60 s (Le Clair & Riach, 1996; Corbeil et al., 2003). The first 10 s of the collected signal were removed to eliminate postural changes that might occur while participants closed their eyes while the last 5 s were removed to eliminate any potential effects of anticipation towards the end of the trial.

The mean of three pre-fatigue sway trials was obtained for different sway measures to improve intra-session reliability (Lafond et al., 2004). Triaxial ground reaction forces and moments were sampled at 100 Hz. The raw signal was low-pass filtered (Butterworth, 5 Hz cut-off frequency, 2^{nd} order, zero lag) and transformed to obtain COP values (Winter 1995). Fast Fourier Transform (FFT) was used to calculate the power spectra for frequencies below 5Hz. In order to implement the FFT, the COP signals were zero-padded to 8192 samples (Loughlin & Redfern, 2001). LabVIEW v6.1 (National Instruments Corporation, Austin, Texas, USA) was used for data collection from the force platform and MatLAB v6.5 (Mathworks, Inc., Natick, Massachusetts, USA) was used for data processing. Joint positions were collected using a passive marker recording system at 20 Hz, Vicon 460 (ViconPeak Motion Systems Inc., Lake Forest, CA) and low-pass filtered (Butterworth, 1.5 Hz cut-off frequency, 2^{nd} order, zero lag). A 13 segment model was used to estimate the whole body COM. Table 3 presents the parameters used to estimate the segmental COM and whole body COM based on a generic model (de Leva, 1996).

Table 3. Whole-body center of mass (COM) model

Segment	Endpoints		Longitudinal COM location (% from proximal endpoint)		Mass (% of body mass)	
	Proximal	Distal	Males	Females	Males	Females
Foot	Lateral Maleolus	5th Metatarsal	44.15	40.14	1.37	1.29
Shank	Knee Joint Center	Lateral Maleolus	44.59	44.16	4.33	4.81
Thigh	Illiospinale (ASIS)	Knee Joint Center	47.63	50.04	14.16	14.78
Torso	Suprasternale	Mid point of the two ASISs	50.96	50.64	43.46	42.57
Head	Mid point of two temple markers	Chin marker (Gonion)	20.49	17.96	6.94	6.68
Upperarm	Acromion	Elbow joint center	62.34	62.17	2.71	2.55
Forearm	Elbow joint center	Wrist joint center	45.74	45.59	1.62	1.38

EXPERIMENTAL DESIGN

The goal of the study was to investigate the application of circumferential ankle pressure (CAP) in mitigating the effects of fatigue on postural sway. A mixed factor design was used, with participants nested within a concomitant variable as described below. Each participant completed all combinations of the independent variables (CAP and fatigue). CAP consisted of two levels (with and without), fatigue also had two levels (before and after).

Blocking variables

Blocking variables used in the current study were PA (low and high) and gender. PA was assessed by evaluating absolute joint position error, and was measured for each participant prior to administering treatments. Joint position sense error was recorded and a PA score was determined accordingly. PA was later divided into two subcategories; namely, low, and high, using a critical joint position sense error of 4.5° (low PA group > 4.5°; high PA group ≤ 4.5°).

The critical joint position sense error was chosen carefully to subdivide participants equally in both groups and at the same time was similar to values suggested in the literature (Perlau, et al., 1995; Callaghan, et al., 2002).

Joint position sense error was assessed using a passive-active joint position sense test protocol (Feuerbach et al., 1994; Voight et al., 1996; Halseth et al., 2004; Miura et al., 2004) during a practice session conducted prior to the exercise sessions. Participants were required to sit on the dynamometer and place their feet on a platform (Figure 4). The platform consisted of a moveable footplate, on which a digital protractor (Pro 360, Kell-Strom, Wethersfield, Connecticut, USA) was mounted. The digital protractor had an accuracy of ± 0.01°. Participants were required to place their dominant foot in a neutral posture on this footplate with their eyes closed and knee adjusted at 90° (Figure 5). Participants' ankles were aligned with the rotational axis of the footplate. After this, the participant's foot was extended passively to 15° plantar-flexion, as this was the midrange of motion for the fatiguing exercise. This position was maintained for 10 s and participants were required to concentrate on this position (Voight et al., 1996). The foot was then returned back to the starting position passively. After this, participants were required to actively reposition their foot to the same angle. A total of three assessments were conducted, each beginning from the starting position. PA was assessed using absolute joint position sense error, calculated as the average angular error (AAE) obtained from the three trials.

Figure 4. Equipment for measuring proprioceptive acuity

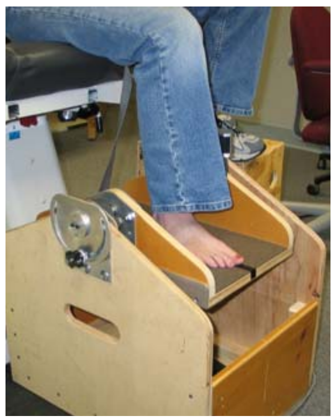

Figure 5. Placement of foot at the center of the footplate and alignment of the ankle joint with the rotational axis of the equipment to measure PA

It is evident that only 1 male and 1 female could be categorized into the low PA and high PA group, respectively (Table 4). The bias inherent in the distribution of PA has potential confounding implications on the experimental design if both, gender and PA are to be considered as separate blocking variables. To eliminate these confounding effects, only gender is considered as the blocking variable in the design.

Table 4. Gender, AAE, and PA (with 4.5° critical AAE) for each participant

Participant No.	Gender	AAE	PA
1	Male	2.83	High
2	Male	4.27	High
3	Male	2.7	High
4	Male	1.77	High
5	Male	1.87	High
6	Male	2.77	High
7	Male	14.57	**Low**
8	Female	7.13	Low
9	Female	6	Low
10	Female	2.8	**High**
11	Female	4.6	Low
12	Female	11.3	Low
13	Female	5.76	Low
14	Female	7.56	Low

Dependent variables

Objective measures

COP based measures

COP based sway measures, which are COP trajectories collected during sway trials, are derived from stabilograms. The stabilograms were used to calculate the following COP based measures.

Ellipse area (EA) is the area of the 95% bivariate confidence ellipse (Prieto et al., 1996).

Mean power frequency (MPF) is the summary measure of the power spectrum (2). Here f is the frequency and P(f) is the power at that frequency (Hasan et al., 1996).

$$MPF = \frac{\sum_{f=0}^{f_n} fP(f)}{\sum_{f=0}^{f_n} P(f)} \qquad (2)$$

Median power frequency (MDPF) is the frequency below which 50% of the total power of the COP spectrum can be found (3), where P is the power spectral frequency, and μ_0 is the total power (Prieto et al., 1996).

$$\sum_{i=1}^{N} P[i] \geq 0.5\mu_0 \qquad (3)$$

Mean velocity (MVELO) of the COP path is calculated for resultant (R), medio-lateral (ML) and antero-posterior (AP) directions as

$$\sum_{i=1}^{N-1} [(x(i+1)-x(i))^2 + (y(i+1)-y(i))^2]^{1/2} / T \qquad (4)$$

where x and y are time series in ML and AP directions, respectively and T is the trial duration

Peak velocity (PVELO) is the maximum velocity and was calculated in R, ML and AP directions.

RMS distance (RDIST) measures the magnitude of COP according to and was used to calculate RMS distance in R, ML and AP directions

$$RDIST = [1/N \sum | R(i) |^2]^{1/2} \qquad (5)$$

where, R is the resultant distance time series obtained from the AP and ML series

Sway area (SA) is used to calculate the area enclosed by COP path per unit of time. This measure can be approximated by adding the area of triangles formed by two consecutive points and the mean COP (6).

$$SA = \frac{1}{2T}\sum_{n=1}^{N-1} | y(n+1) * x(n) - y(n) * x(n+1) | \tag{6}$$

where, T is the trial duration; x and y are the time series in ML and AP directions, respectively and N is the number of samples

Detrended fluctuation analysis (DFA) is a fractal analysis approach to calculate the scaling exponent. DFA was developed specifically to analyze biological time series such as these that are highly non-stationary (Peng et al., 1995) and is calculated by first subtracting the mean of the data set, x_m, from the data series x(i) and then by integrating the time series (7a).

$$y(k) = \sum_{i=1}^{k} (x(i) - x_m) \tag{7a}$$

The resulting time series integrated signal is then subdivided into equal non-overlapping intervals of length n. There is no specific standard on the choice of interval length, but typical suggested ranges are from 3 samples to $1/4^{th}$ the length of series (Popinov & Minerva, 1999), and 10 to ½ the length of series (Norris et al., 2005). The purpose is to identify whether the chosen signal can be illustrated as a self-similar process after integration. If the accumulated sum (obtained after integrating the signal) can be scaled as a power law (with different window sizes), it is considered self-similar (Hausdorff et al. 1996). For the current study, the interval chosen was 10 to ½ the length of the series.

The final step in the process of DFA is to determine a locally (within each window) best fit line, which is obtained by performing a linear least-squares fit. This procedure yields what is termed a local trend, ŷ (k). The integrated signal is then detrended by subtracting the local trend within each window (or interval). A RMS of the integrated and detrended time series signal is then calculated (7b).

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}[y(k) - \hat{y}(k)]^2} \tag{7b}$$

The RMS fluctuation of the integrated and detrended time series may exhibit a scaling law relationship with the above chosen interval length. This is where DFA bears its resemblance to the Hurst rescaled adjusted range analysis (HRARA) or R/S analysis or statistic. The R/S statistic was proposed by Hurst and has been commonly used in the past to demonstrate and describe the fractal properties of COP signals collected over time (Duarte & Zatsiorsky, 2000). The scaling law relationship is expressed according to

$$F(n) = \propto n^{\alpha} \tag{7c}$$

where the scaling exponent, α varies between 0 and 1.5.

If the signal behaves like an uncorrelated random walk process, commonly known as white noise, the scaling exponent $\alpha = 0.5$, whereas if the signal behaves as a *Fractional Brownian motion*, $\alpha = 1.5$. Delignieres et al. (2003) demonstrated that both scaling exponents α and $H_{R/S}$ obtained from DFA and HRARA, respectively, are related to the scaling exponent H derived from Stabilogram Diffusion Analysis (SDA, Collins & DeLuca, 1993), as

$$H_{R/S} = H + 0.5$$

$$\alpha = 2 * H - 0.5 \tag{7d}$$

Since the scaling exponents obtained from HRARA and SDA have in the past been shown to be related with α, the other two methods of analysis were not used in the current report. In the current study, α (DFAEX), the power scaling exponent obtained from DFA, was used as a dependent measure. DFA in both ML and AP directions was used to analyze stationarity of the COP signals.

Previous studies have reported contradictory findings in the stationarity of COP signal using different fractal analysis methods, (Duarte & Zatsiorsky, 2000; Ferdjallah, et al., 1999; and Schumann, et al., 1995). Thus, the current study used DFA to investigate the stationarity of COP signal. In addition to the scaling exponent obtained from DFA (α), another scaling exponent, (β) was also used. β is the negative slope of the locally best fit line for a log-log plot of power and frequency. For infinite stationary time series, the two scaling exponents are theoretically related as (Hausdorf et al., 1995; and Hausdorf et al., 1996)

$$\alpha = (\beta + 1)/2 \qquad (7e)$$

Thus, to validate the stationarity of COP signals in ML and AP the scaling exponent α, obtained using the DFA, the DFAEX, was compared to the scaling exponent, β (7e). For the purposes of statistical analysis the difference between the two exponents was obtained.

Mean COP position (MCOP) in ML and AP directions was calculated to analyze characteristics of postural adaptation after being exposed to localized muscle fatigue.

Modified Time to boundary (MTtb) has been derived and modified from the *time-to-contact* between COP trajectory and stability boundaries which is also known as time-to-boundary or Ttb (Wegen et al. 2002). Riccio (1993) suggested that the effort required in maintaining balance (compensatory torque) is influenced by the direction of balance. An individual's perception of spatio-temporal proximity to the limits for maintaining balance is commonly known as the stability boundaries (SBs). If the spatio-temporal proximity to the limits were less, it would fundamentally minimize the time-to-contact with the SBs. In a situation where the time-to-contact was less than the time required for a particular action/task system (e.g. quiet standing), another action (e.g. stepping) would have to be chosen to prevent the onset of disequilibrium. Wegen et al. (2001) modified the concept of *time-to-contact* in order to better investigate the relation of COP movement to the SBs (defined as Ttb, Wegen et al. 2002). The SBs were determined by circumscribing a rectangle around the physical dimensions of BOS. The instantaneous distance (d) and velocity (V_{COP}) with respect to the SB were then used to calculate Ttb, as:

$$Ttb(i) = d(i) / V_{COP}(i) \tag{8}$$

Ttb was modified here to obtain Modified Time-to-boundary (MTtb). In order to closely encompass the BOS an irregularly shaped octagon was used. The octagon represented the contour of the BOS as tightly as possible and was standardized for all conditions to obtain MTtb (8). The COP position (stabilograms) and velocity (with arrows) vectors were obtained to illustrate the BOS boundary with which the COP position vector at a given instant would eventually collide (Figures 6 & 7). During quiet stance, the COP trajectory never actually collides with the BOS (or SBs), and thus the MTtb can never be reliably assessed at the point of maximum excursion (i.e. when velocity tends to zero). To avoid any inconsistencies, only COP signal with velocities below the 95th %-ile were used for MTtb calculations. The current study uses mean, standard deviation and minimum MTtb (MMTtb, STDEVTtb, MinMTtb, respectively) as the dependent measures.

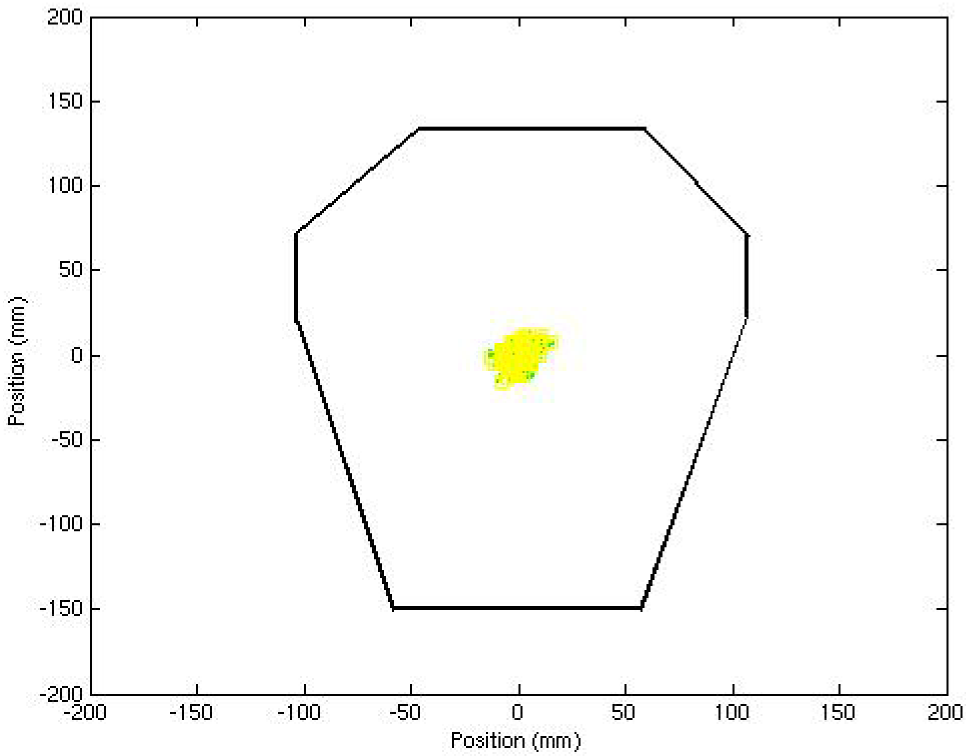

Figure 6. BOS and COP position and velocity vectors

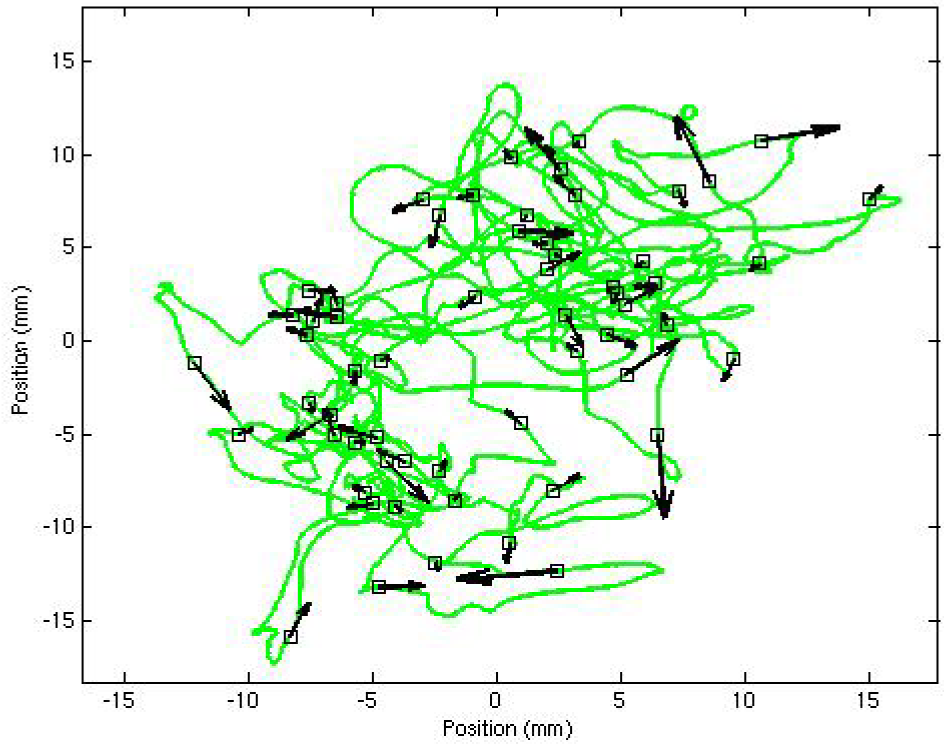

Figure 7. COP position and velocity vectors at approximately the center of the BOS

Kinematics based measures

RMS of ankle and hip angles (RMSAnTh and RMSHiTh) were used as dependent measures in an attempt to investigate different strategies as well as postures attained, both before and after, the onset of localized muscle fatigue and/or application of CAP.

Theta (θ; Th) the covariance of two body angles, ankle and hip, was used in the current study in order to investigate postural coordination. Previous studies have described two different strategies exercised during neuromuscular control and maintenance of quiet standing, namely: "ankle strategy" and "hip strategy" (Nashner & McCollum, 1985). *Th* was used to demonstrate postural coordination, if any, between these two different strategies. Kuo et al., (1998) used shank and hip angles to derive a positive definite variance and covariance matrix to describe postural coordination. To better investigate the phase relationship between movements around the ankle and hip joints the phase plot used the ankle instead of shank angle as the abscissa.

Combined COP and COM measures

COP-COM signal (ERR) has been used in the past to overcome some of the limitations that intrinsically exist when COP and COM based measures are considered separately to describe the postural control (Corriveau et al., 2001). The COP-COM variable is the instantaneous scalar distance between the COP and COM. The COP-COM amplitude is not only proportional to the horizontal acceleration of COM during quiet and unperturbed stance, but it has been used to demonstrate the efficacy of postural control (Winter, 1995). The current study uses RMS and Peak-to-Peak amplitude of the COP-COM signal (RMSERR and PtPERR, respectively) in both ML and AP directions. Whole body COM estimates in the current study contain a bias error as it is calculated from a generic model (de Leva, 1996). During quiet standing this bias error will be constant throughout the duration of the trial. To avoid such bias, the mean COM was subtracted from the COM signal (Winter et al., 1998), according to

$$COM(i) = COM(i) - \overline{COM} \tag{9}$$

Stiffness (STIFF) in ML and AP directions was estimated using a direct estimation method (Winter et al., 2001). If COP and COM both are measured with respect to the ankle joint, assuming that COM of the body is located *h* distance above the ankle joint, then

$$M_a = R * COP = mg * COP \quad (10a)$$

$$\theta_{sw} = \frac{COM}{h} \quad (10b)$$

$$Stiffness(K_a) = \frac{dM_a}{d\theta_{sw}} \quad (10c)$$

where M_a is the sum of left and right ankle moments, θ_{sw} is the sway angle determined by the angle of line joining the ankle to the COM, *mg* is the body weight above the ankle joint, *R* is the vertical reaction force, and h is height of the COM from the ankle joint.

Subjective measures

Perceived Stability Rating (PSR) was obtained at the end of each sway trial using the scale described above. Participants were asked to evaluate their stability by perceiving how stable/steady they felt while performing the quiet standing trial (Schieppati, et al., 1999).

STATISTICAL ANALYSES

Stationarity test for COP signals

One-sample t-tests were performed on mean pre-fatigue and post-fatigue *DFAEX* ML and AP. Scaling exponents, β, were calculated for the COP signals in ML and AP and the DFAEX ML and AP were compared to α (6e). The difference between the two scaling exponents *DFAEX* and β was obtained. The mean of this difference was tested against a test mean of 0 with significance at $p < 0.05$.

Correlation matrix for Dependent measures

Previous studies have shown high correlations among COP based sway measures (Prieto et al. 1996). A correlation analysis was performed on all dependent measures and Pearson's correlation values greater than the critical correlation value, were selected. The critical correlation value was determined based on the sample size of 14 (number of participants) and $\alpha = 0.05$.

Effects of CAP and fatigue on postural control

A randomized block design with blocks formed from PA was used to analyze all dependent measures used in the study, for the effects of CAP (with and without) and fatigue (pre- and post-). Additionally, due to the inherent bias in the PA categorization and gender all the dependent measures were analyzed separately with gender as the blocking variable. The statistical model for the design is:

$$Y_{ijklm} = \mu_{....} + \rho_i + \alpha_j + \beta_k + \rho\alpha_{ij} + \rho\beta_{ik} + \alpha\beta_{jk} + \gamma_l(\rho_i) + \rho\alpha\beta_{ijk} + \varepsilon_{ijklm} \qquad (11)$$

where: Y_{ijklm} are the dependent sway measures;

$\mu_{....}$ is the overall mean,

ρ_i are fixed effects, subject to restriction $\Sigma\, \rho_i = 0$ and represents blocking variable PA or gender,

α_j and β_k are fixed effects, representing factors CAP and fatigue respectively, subject to restrictions $\Sigma\, \alpha_j = \Sigma\, \beta_k = 0$,

γ_l are the random effect representing participants and are independent with N $(0, \sigma^2_\gamma)$

ε_{ijklm} are the error terms and are independent with with N $(0, \sigma^2)$

Separate ANOVAs were performed on each dependent measure to analyze the effects of CAP and fatigue on dependent sway measures. A p value of 0.1 was used to determine significance of effects for all statistical analyses. Due to deviation from normality, logarithmic transformations were performed on Ellipse area, Median frequency AP, Peak velocity, RMS distance, RMS distance ML, Sway area and Peak-to-Peak COP-COM ML.

RESULTS

STATIONARITY TEST FOR COP SIGNAL

Table 5 presents mean differences between DFAEX and theoretical α (6e), in the ML and AP directions, across all participants trials. The non-significant difference between mean pre-fatigue DFAEX and theoretical α (Table 5) indicates that COP signals were predominantly stationary, except mean pre-fatigue COP signals in AP.

Table 5. Mean difference between DFAEX and theoretical α, *p*-values from t-test in parentheses

	ML	**AP**
Mean pre-fatigue	0.0014 (0.9502)	**-0.0582** (0.0002)
Mean post-fatigue	0.0048 (0.8566)	-0.034 (0.1805)

CORRELATION ANALYSIS FOR DEPENDENT MEASURES

Correlation values were obtained for the dependent measures. The critical value of Pearson's correlation coefficient for the sample size of 14 and $\alpha = 0.05$, is $r = 0.458$. However, in order to highlight relatively high values, only those with Pearson's correlation, $r > 0.6$ are presented (Table 6). The velocity based measures were highly correlated among each other, as well as with the ellipse area, sway area, and negatively correlated with time-to-boundary measures (Table 6). Peak-to-Peak COP – COM signal was also found to be highly correlated with velocity measures. It is interesting to note that postural steadiness ratings were negatively correlated with velocity based measures. The frequency based measures were correlated with the DFA scaling exponent.

There was negligible correlation found between mean COP position and stiffness measures and other COP based measures. Additionally, the Kinematics based measures were correlated among each other but not with the COP based measures, with the exception of Peak-to-Peak COP-COM signal, which was, interestingly, correlated with mean velocity measures.

Table 6. Matrix for different dependent measures and correlates (r > 0.6 between measures in the first column, and all those in the corresponding row)

Measures						Correlates						
EA	MVELO AP	PVELO AP	RDIST	RDIST ML	RDIST AP	SA						
MPF ML	MDPF ML	DFAEX ML										
MPF AP	MDPF AP	DFAEX AP										
MDPF ML	MPF ML	DFAEX ML										
MDPF AP	MPF AP	DFAEX AP										
MVELO	MVELO ML	MVELO AP	PVELO	PVELO ML	PVELO AP	SA	MMTtb	MinMTtb	PtPERR AP	PSR		
MVELO ML	MVELO AP	PVELO	PVELO ML	PVELO AP	SA	MMTtb	MinTtb	PtPERR ML	PSR			
MVELO AP	EA	PVELO	PVELO ML	PVELO AP	RDIST	RDIST ML	RDIST AP	SA	MMTtb	MinMTtb	PtPERR AP	PSR
PVELO	MVELO	MVELO ML	MVELO AP	PVELO ML	PVELO AP	SA	MMTtb	MinMTtb	PtPERR AP	PSR		
PVELO ML	MVELO	MVELO ML	MVELO AP	PVELO	PVELO AP	SA	MMTtb	MinMTtb				
PVELO AP	MVELO	MVELO ML	MVELO AP	RDIST	RDIST ML	SA	MMTtb	MinMTtb	PtPERR AP			
RDIST	EA	MVELO AP	RDIST ML	RDIST AP	SA							
RDIST ML	EA	MVELO AP	PVELO AP	RDIST	RDIST AP	SA						
RDIST AP	EA	MVELO AP	RDIST	RDIST ML	SA							
SA	EA	MVELO	MVELO ML	MVELO AP	PVELO	PVELO ML	PVELO AP	RDIST	RDIST ML	RDIST AP		
DFAEX ML	MPF ML	MDPF ML										
DFAEX AP	MPF AP	MDPF AP										
MCOP ML												
MCOP AP												
MMTtb	MVELO	MVELO ML	MVELO AP	PVELO	PVELO ML	PVELO AP	MinTtb					
STDEVMTtb												
MinMTtb	MVELO	MVELO ML	MVELO AP	PVELO	PVELO ML	PVELO AP						
RMSAnTh ML	RMSAnTh AP											
RMSAnTh AP	RMSAnTh ML											
RMSHiTh ML	Th ML	Th AP										
RMSHiTh AP	Th AP											
Th ML	RMSHiTh ML											
Th AP	RMSHiTh ML	RMSHiTh AP										
RMSERR ML												
RMSERR AP												
PtPERR ML	MVELO ML											
PtPERR AP	MVELO	MVELO AP	PVELO	PVELO AP								
STIFF ML	STIFF ML											
STIFF AP	STIFF AP											
PSR	MVELO	MVELO ML	MVELO AP	PVELO								

EFFECTS OF CAP, FATIGUE AND GENDER ON POSTURAL CONTROL

PA as a blocking variable

Table 7 presents the overall results obtained from ANOVAs performed on all the dependent measures with PA as the blocking variable.

Table 7. p-values for main and interaction effects of CAP, fatigue and gender on all dependent measures with PA as blocking variable

Dependent Measures	Factors						
	CAP	**Fatigue**	**PA**	**CAP X Fatigue**	**CAP X PA**	**Fatigue X PA**	**CAP X Fatigue X PA**
EA	0.2199	0.2334	0.4982	0.7744	0.1660	0.8566	0.4144
MPF ML	0.2062	0.4472	0.3475	0.1916	0.2373	0.8119	0.9695
MPF AP	0.4210	0.6900	0.9159	0.3345	0.7453	0.8644	0.8075
MDPF ML	0.1789	0.6538	0.4157	0.3632	0.1566	0.9118	0.8649
MDPF AP	0.8287	0.9860	0.3454	0.3497	0.5357	0.9426	0.3014
MVELO	0.4412	**0.0019**	0.2680	0.2635	0.2330	0.3458	0.2475
MVELO ML	0.4534	**0.0035**	0.3398	0.1751	0.4613	0.3317	0.2427
MVELO AP	0.5042	**0.0030**	0.2236	0.5656	**0.0848**	0.4161	0.3238
PVELO	0.2908	0.6233	0.2839	0.6776	**0.0784**	0.4646	0.1705
PVELO ML	0.1859	0.1277	0.4346	0.2517	0.1173	0.4846	0.3467
PVELO AP	0.6577	**0.0963**	0.4162	0.9089	**0.0867**	0.8297	0.1173
RDIST	0.1955	0.2963	0.5317	0.8361	0.2572	0.7612	0.4471
RDIST ML	0.1806	0.3112	0.7721	0.7532	0.2433	0.5201	0.5138
RDIST AP	0.2914	0.3164	0.3257	0.3167	0.2266	0.6857	0.2806
SA	0.2658	**0.0377**	0.2715	0.5373	0.3007	0.3626	0.3144
DFAEX ML	**0.0954**	0.7939	0.5330	0.3806	0.3256	0.5096	0.1780
DFAEX AP	0.6044	0.6931	0.7958	0.9350	0.8910	0.9885	0.6606
MCOP ML	0.2409	**0.0049**	0.3797	0.8210	0.3854	0.5968	0.5487
MCOP AP	0.6938	**0.0177**	0.8942	0.6397	0.6404	0.8307	0.4003
MMTtb	0.4069	**0.0168**	0.2136	0.2237	0.9797	0.2445	0.9163
STDEVTtb	0.5719	0.9889	0.3941	0.1704	**0.0661**	0.8884	**0.0059**
MinMTtb	0.9273	0.3055	0.2876	0.6469	0.2905	0.9157	0.1048
RMSAnTh ML	0.4354	0.5415	0.8392	0.6248	0.1550	0.7687	0.9589
RMSAnTh AP	**0.0054**	0.1244	0.4110	0.3431	0.7535	0.7819	0.8590
RMSHiTh ML	**0.0138**	0.5105	0.8634	0.6973	**0.0039**	0.4964	0.8625
RMSHiTh AP	0.1502	0.2100	0.8059	0.8361	0.3483	0.7806	0.6142
Th ML	0.9824	0.3955	0.9416	0.7345	0.2796	0.2062	0.7061
Th AP	0.8374	0.6103	0.7551	0.7056	**0.0670**	0.8721	0.2008
RMSERR ML	**0.0014**	0.9099	0.0787	0.7752	0.5543	0.4410	0.7266
RMSERR AP	**0.0793**	0.4711	**0.0917**	0.6443	0.4930	0.3203	0.5090
PtPERR ML	0.2448	0.7005	**0.0495**	0.6999	0.3959	0.4671	0.7306
PtPERR AP	0.5155	0.6132	0.4820	0.3398	**0.0127**	0.7917	0.3783
STIFF ML	0.7958	0.1712	**0.0165**	0.2966	0.5168	0.2313	0.6961
STIFF AP	0.6275	**0.0959**	**0.0429**	0.9279	0.9994	0.8059	0.1145
PSR	0.7570	0.1043	0.8278	0.6164	0.6853	0.2052	0.6697

p-values < 0.1 in bold

COP based measure DFA scaling exponent in ML, kinematics based measures like Root Mean Square of the ankle angle in AP and hip angle in ML, and the angle formed by the phase angle plot of hip vs. ankle angles demonstrated the main effects of CAP (Table 7).

COP based measures mean velocity, peak velocity, sway area, and mean mod. Time-to-boundary significantly illustrated the main effects of fatigue. Interestingly, the mean COP position in ML as well as AP was affected by fatigue. Finally, stiffness in AP and postural stability ratings also demonstrated main effects of fatigue (Table 7). There was approximately a 12%, 26% and 46% change in mean sway velocity, sway area and mean COP position in ML, respectively, post-fatigue (Table 8).

Table 8. Percent changes in dependent measures post-fatigue

Dependent Measures	**Fatigue**		**% Change**
	Pre-fatigue	**Post-fatigue**	
Mean velocity (m/s)	0.0145	0.0163	12.41
Sway area	3.62 e-5	4.56 e-5	25.97
Mod. Mean time-to-boundary (s)	9.1078	8.3356	8.48
Mean COP ML (m)	-0.0063	-0.0092	46.03
Mean COP AP (m)	0.0466	0.0513	10.09
Stiffness AP	557.97	578.58	3.69

Hypothesis 1: CAP will be more effective in reducing postural sway in individuals with low PA during upright stance

The interaction of CAP and PA, affected mean velocity in AP, peak velocity, peak velocity in AP, standard deviation of mod. time-to-boundary measure, root mean square of the hip angle in ML, angle formed by the phase plot of the hip vs. ankle angle in AP, and peak-to-peak amplitude of the COP-COM in AP. Figures 8 & 9, illustrate the interaction effect of CAP and PA on mean velocity in AP (m/s) and standard deviation of mod. Time-to-boundary (s). The application of CAP reduced the mean velocity of sway in AP in participants with high PA, whereas in participants with low PA the application of CAP increased the mean velocity of sway in AP (Figure 8). The mean velocity of sway decreased about 9% with the application of CAP in individuals with higher PA, whereas it increased by about 6% in individuals with lower PA. Similarly, the application of CAP reduced the variability in time-to-boundary in high PA group, whereas the variability in time-to-boundary increased the variability in time-to-boundary for individuals with low PA (Figure 9). The application of CAP increased the variability in time-to-boundary by about 2.3% in individuals with low PA, but decreased it by 1.5% in individuals with high PA. To summarize, the effects of CAP on postural control, although significant, were minor for individuals with both low as well as high PA.

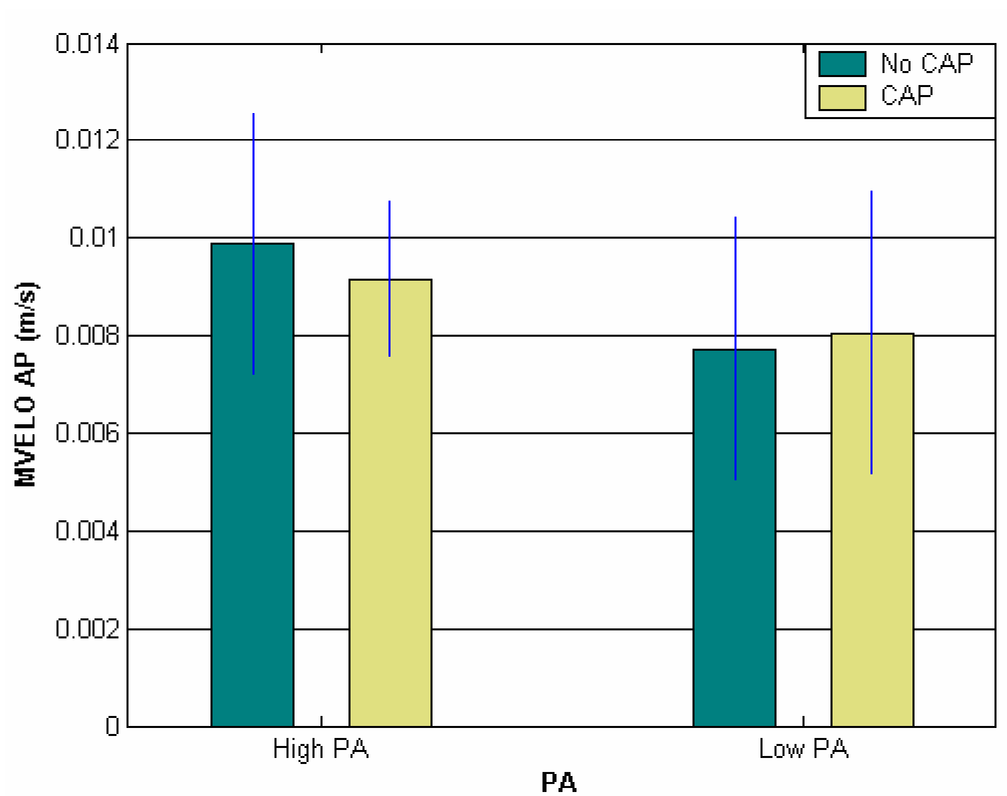

Figure 8. Interaction effects of CAP and PA on Mean Velocity AP

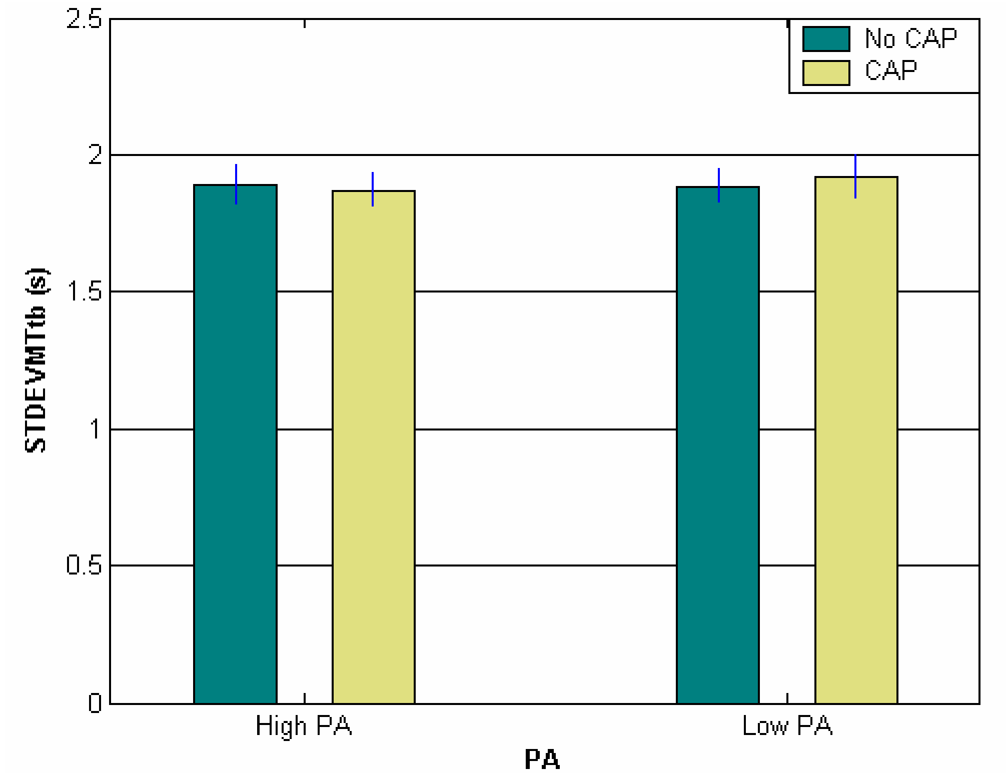

Figure 9. Interaction effects of CAP and PA on Standard deviation of Mod. Time-to-boundary

Hypothesis 2: Fatigue effects in individuals with low PA will be more severe as compared to others during upright stance

The interaction of fatigue and PA did not have significant effect on any of the dependent measures that were used in the study.

Hypothesis 3: CAP will be more effective in maintaining postural stability after it has been perturbed by the effect of localized muscle fatigue in individuals with low PA

Of all the 35 dependent measures used, only standard deviation of mod. time-to-boundary illustrated the three-way interaction effects of CAP, fatigue and PA (Table 7). It is evident that the STDEVMTtb AP was smaller after fatigue for low PA individuals without the application of CAP (Figure 10; top) however, it was slightly larger post-fatigue with the application of CAP (Figure 10; bottom). For high PA group with the application of CAP, the variability in time-to-boundary decreased post-fatigue (Figure 10; bottom).

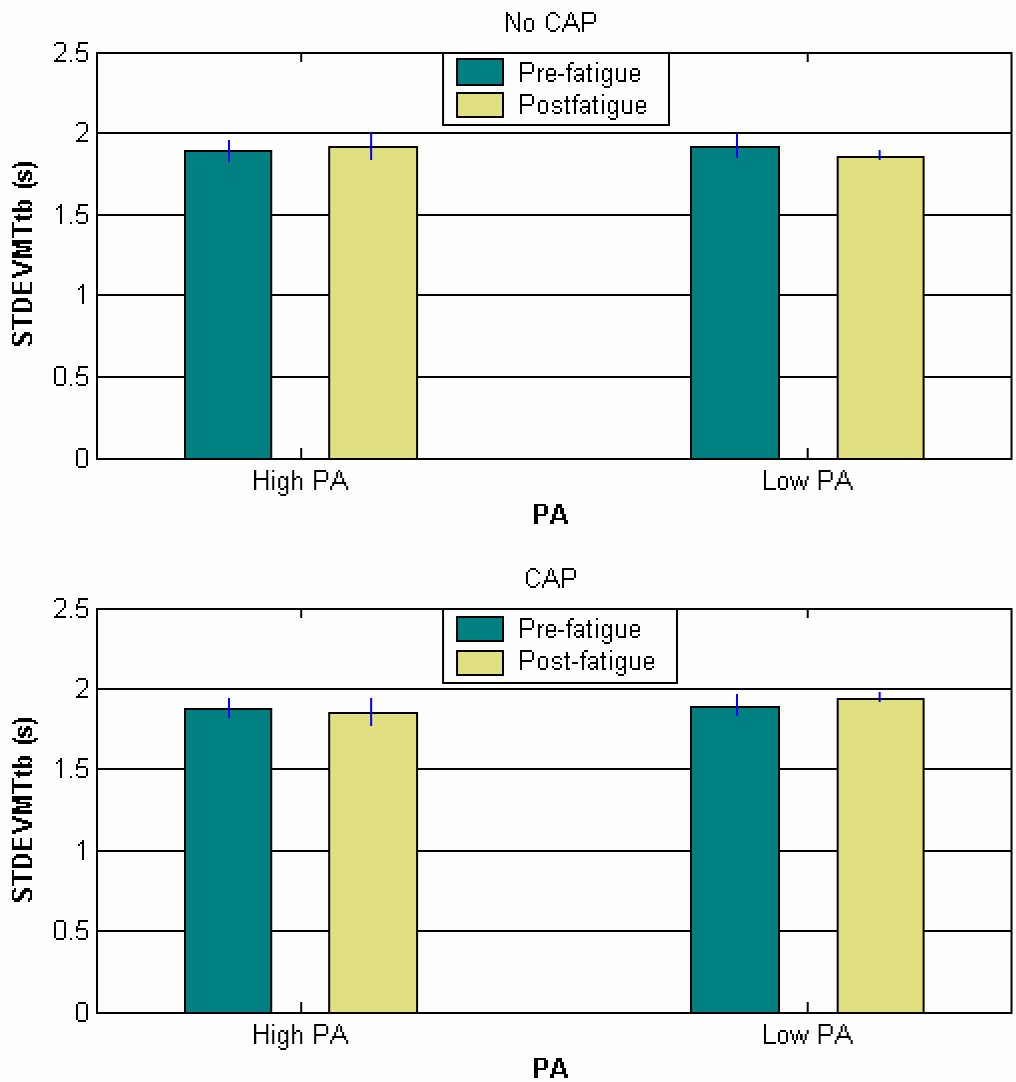

Figure 10. Three-way interaction between CAP, fatigue and PA on Standard deviation of Mod. Time-to-boundary

Gender as a blocking variable

Table 9 presents the overall results obtained from ANOVAs performed on all the dependent measures.

Table 9. p-values for main and interaction effects of CAP, fatigue and gender on all dependent measures

Dependent Measures	**Factors**						
	CAP	**Fatigue**	**Gender**	**CAP X Fatigue**	**CAP X Gender**	**Fatigue X Gender**	**CAP X Fatigue X Gender**
EA	0.197	0.210	0.171	0.763	**0.031**	0.281	0.397
MPF ML	0.193	0.434	0.611	0.179	0.201	0.188	0.573
MPF AP	0.415	0.686	0.255	0.328	0.596	0.359	0.843
MDPF ML	0.165	0.643	0.458	0.347	0.343	**0.068**	0.547
MDPF AP	0.434	0.953	**0.089**	0.378	0.455	0.326	0.668
MVELO	0.419	**0.001**	0.355	0.241	**0.033**	0.383	0.157
MVELO ML	0.433	**0.002**	0.491	0.156	**0.080**	0.303	0.136
MVELO AP	0.490	**0.002**	0.261	0.552	**0.018**	0.658	0.258
PVELO	0.116	0.252	0.537	0.329	**0.005**	0.521	0.185
PVELO ML	0.164	0.109	0.599	0.228	**0.017**	0.429	0.231
PVELO AP	0.642	**0.081**	0.474	0.904	**0.016**	0.550	**0.079**
RDIST	0.173	0.272	0.152	0.828	**0.048**	0.272	0.419
RDIST ML	0.139	0.263	0.202	0.728	**0.010**	**0.086**	0.571
RDIST AP	0.286	0.311	0.190	0.311	0.170	0.923	0.180
SA	0.160	**0.008**	0.288	0.287	**0.024**	0.251	**0.099**
DFAEX ML	**0.098**	0.796	0.443	0.385	0.278	0.369	0.452
DFAEX AP	0.602	0.691	0.251	0.935	0.735	0.849	0.458
MCOP ML	0.225	**0.004**	0.453	0.815	0.419	**0.071**	0.978
MCOP AP	0.692	**0.017**	0.191	0.637	0.670	0.796	0.259
MMTtb	0.359	**0.025**	0.896	0.661	0.276	0.614	0.311
STDEVMTtb	**0.012**	0.548	0.148	**0.084**	**0.087**	0.655	0.165
MinMTtb	0.354	0.885	0.977	0.518	0.566	0.686	0.493
RMSAnTh ML	0.442	0.547	0.138	0.630	0.359	0.688	0.775
RMSAnTh AP	**0.004**	0.114	0.252	0.329	0.148	0.614	0.841
RMSHiTh ML	**0.039**	0.578	0.564	0.742	0.597	0.308	0.941
RMSHiTh AP	0.152	0.180	0.189	0.826	**0.026**	0.850	0.598
Th ML	**0.001**	0.901	0.294	0.780	0.172	0.920	0.669
Th AP	**0.084**	0.476	0.722	0.647	0.846	0.632	0.568
RMSERR ML	0.401	0.280	0.377	0.372	0.157	0.369	0.102
RMSERR AP	0.391	0.582	0.279	0.787	**0.079**	0.433	0.445
PtPERR ML	0.390	0.905	0.166	0.543	0.231	0.619	0.287
PtPERR AP	0.318	0.902	0.488	0.381	**0.079**	0.318	**0.045**
STIFF ML	0.499	0.127	**0.004**	0.202	0.967	0.531	0.754
STIFF AP	0.244	**0.046**	**0.009**	0.448	0.164	0.792	0.313
PSR	0.751	**0.096**	0.716	0.607	0.694	**0.086**	0.395

p-values < 0.1 in bold

CAP and fatigue significantly affected various dependent measures (Table 8) and these are similar to the analysis performed with PA as the blocking variable (Table 7).

The interaction of CAP and gender, affected all the velocity measures, root mean square distance in R and ML directions, sway area, standard deviation of mod. time-to-boundary measure, root mean square of the hip angle in AP, root mean square of the COP-COM in AP, and peak-to-peak amplitude of the COP-COM in AP. The application of CAP reduced the mean velocity of sway in AP among males, whereas in females the application of CAP increased the mean velocity of sway in AP (Figure 11). Similarly, the application of CAP reduced the variability in time-to-boundary in males, whereas the variability in time-to-boundary increased the variability in time-to-boundary for females (Figure 12). Similar to the PA categorization, the effects of CAP on postural control, although significant, were small in both males and females.

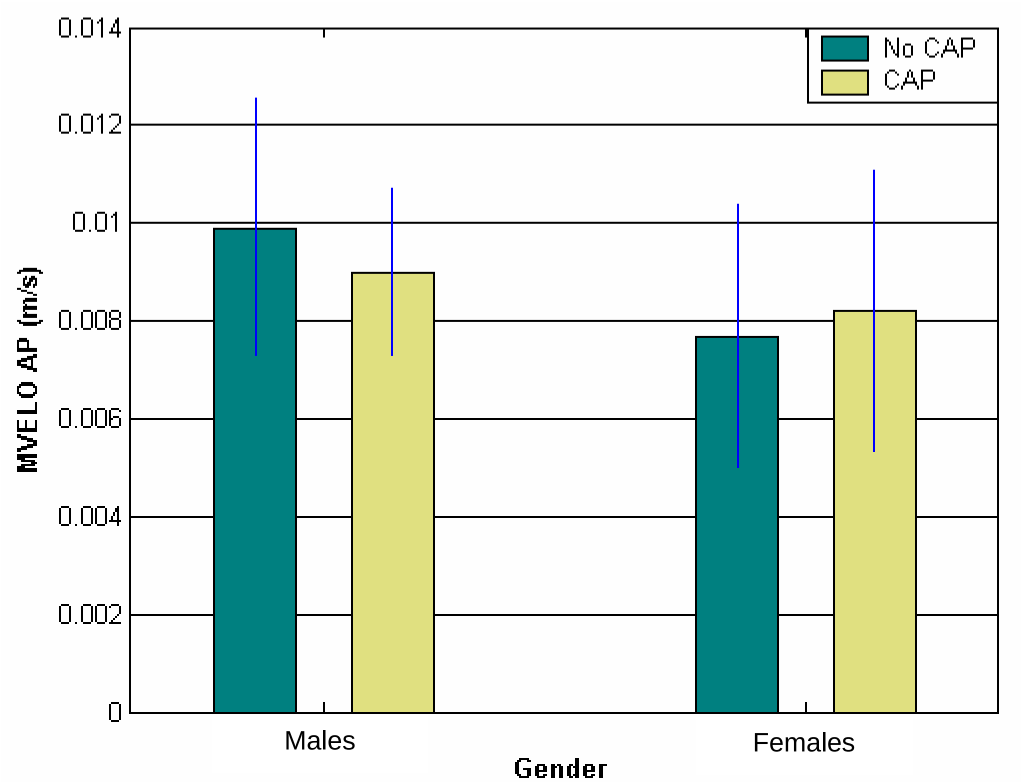

Figure 11. Interaction effects of CAP and gender on Mean Velocity AP

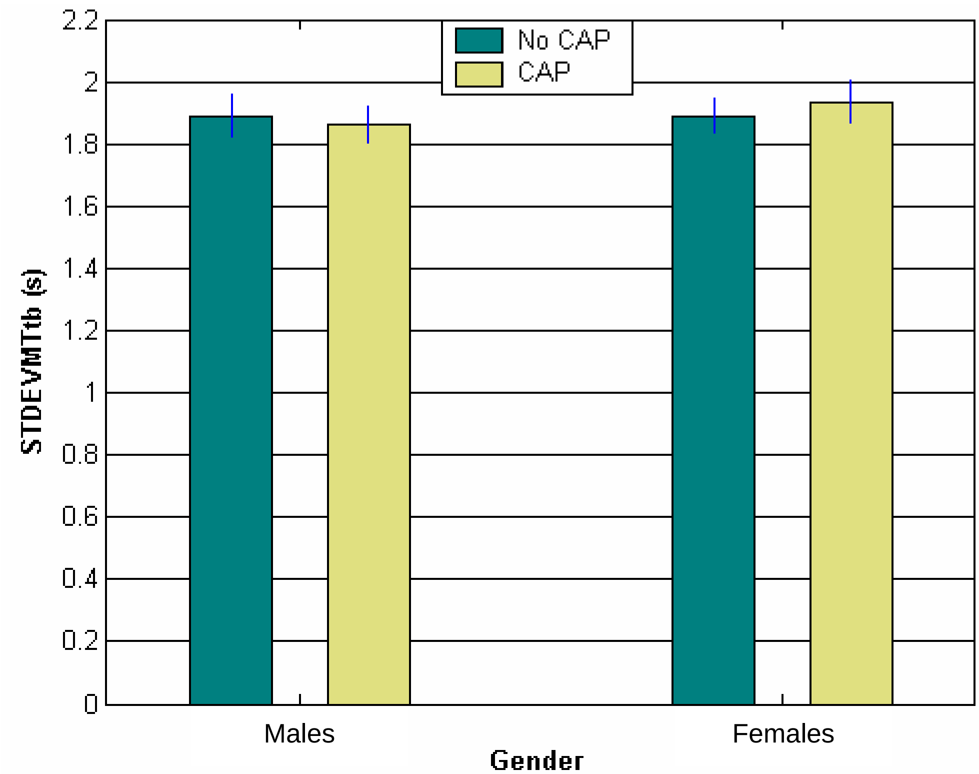

Figure 12. Interaction effects of CAP and gender on Standard deviation of Mod. Time-to-boundary

Interaction of fatigue and gender significantly affected median power frequency in AP, root mean square distance in ML, mean COP position in ML and postural stability ratings. Median power frequency in AP was lower for males ($p < 0.1$) than for females (Figure 13). For females, median frequency in AP increased after the induction of fatigue, whereas it decreased after fatigue for males. In other words, females swayed with increased higher frequency components post fatigue. Females recorded an increase of 11% post-fatigue in median power frequency AP, while males recorded a decrease of 15% post-fatigue.

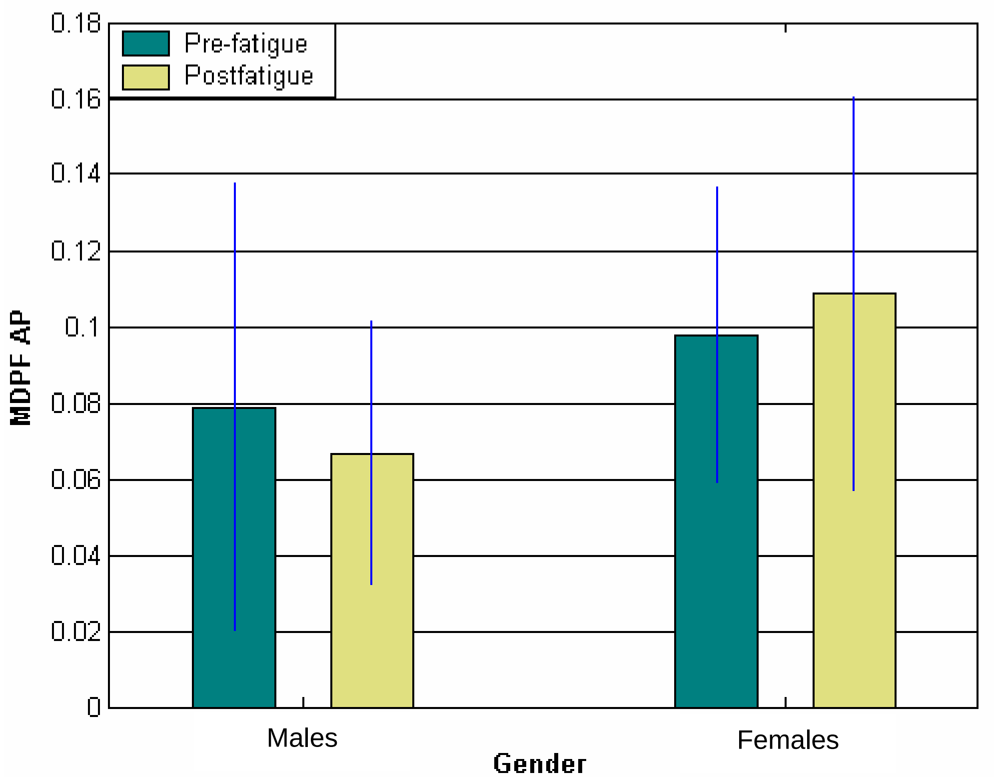

Figure 13. Interaction effects of fatigue and gender on Median power frequency AP

Interaction effects of fatigue and gender on root mean square distance in ML were significant (Table 8). The RMS value of the distance in ML, increased by about 19% for post-fatigue condition in males, whereas RDIST ML remained the same after the induction of fatigue for females (Figure 14).

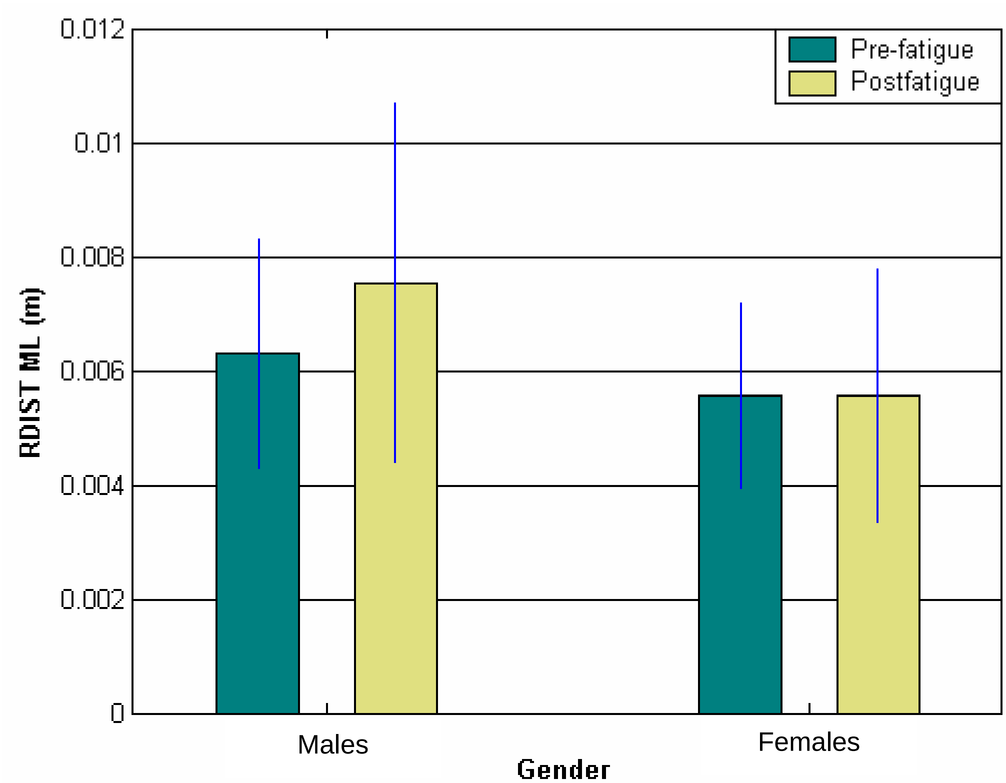

Figure 14. Interaction effects of fatigue and gender on RMS distance in ML direction

The interaction of fatigue and gender significantly affected mean COP position in ML (Table 8). In order to better comprehend the lateral shift from the center of force plate the mean COP position in ML is represented as an absolute value here (Figure 15). Although, both males and females displayed a lateral shift in the mean COP position after fatigue, this lateral shift was less prominent (about 15%) in females than in males, among whom the lateral shift almost increased by 100%. However, it should be noted here that females displayed a greater shift in mean COP position pre-fatigue as compared to males.

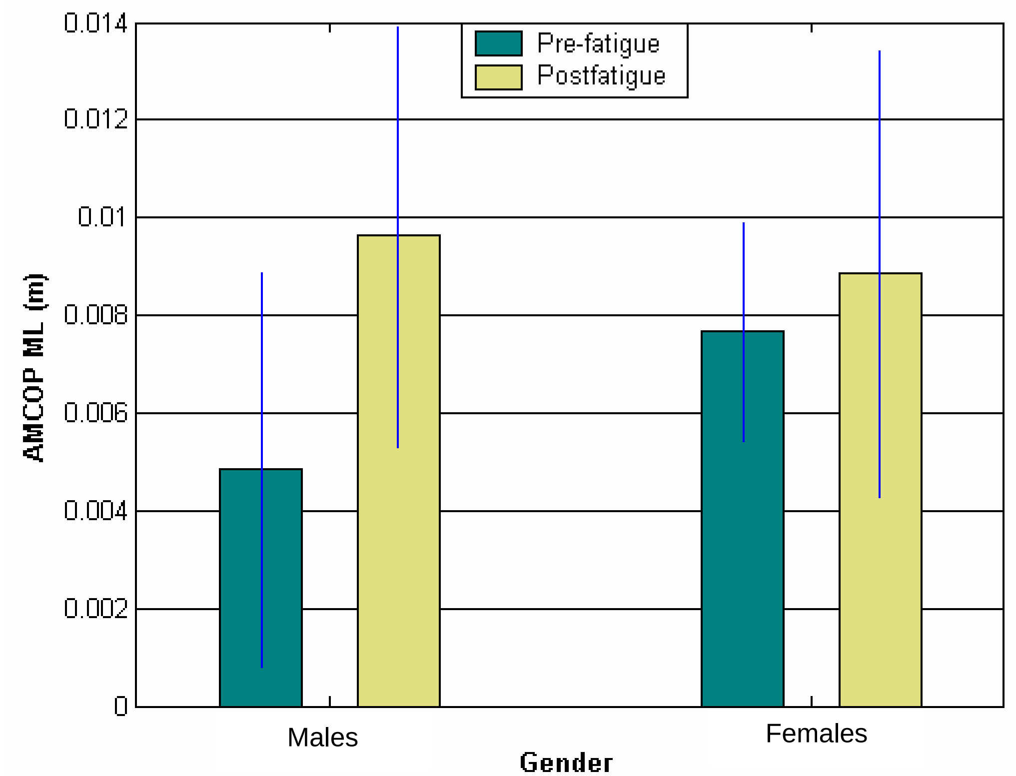

Figure 15. Interaction effects of fatigue and gender on Absolute mean COP position in ML

Perceived stability ratings (PSR) were significantly affected by the interaction of fatigue and gender as well (Table 8). Males assessed their perceived stability better than females after fatigue (Figure 16). It should be noticed here that the pre-fatigue PSR for males was greater than post-fatigue PSR, whereas PSR pre- and post-fatigue remained the same for females. Additionally, males assessed better stability ratings pre-fatigue when compared to females.

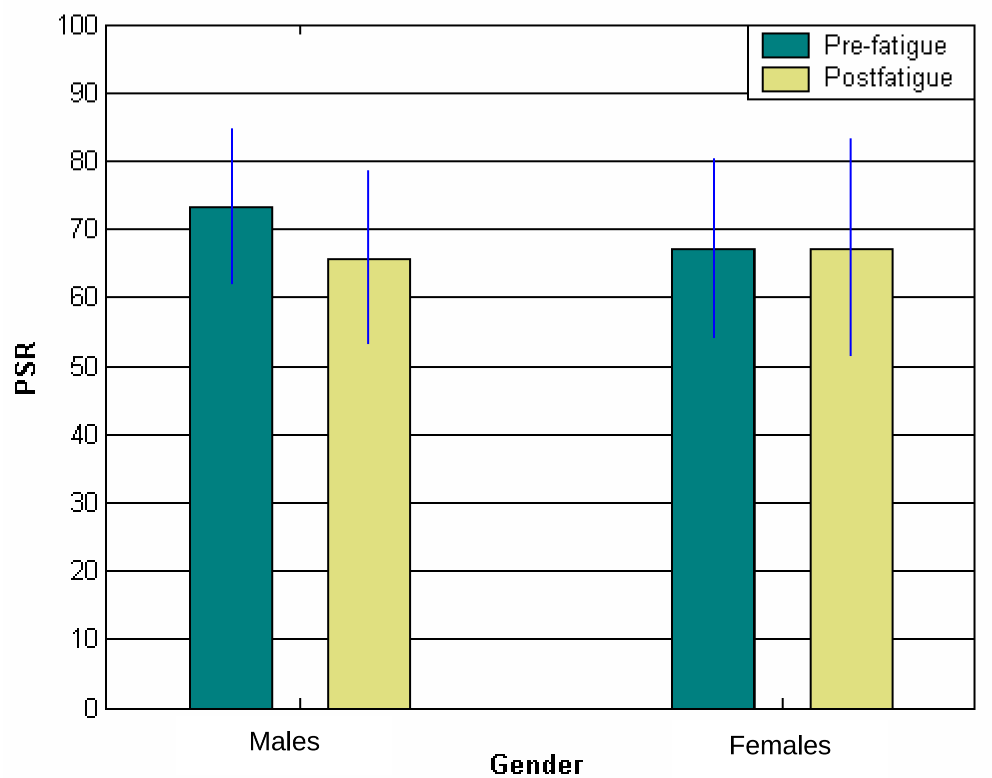

Figure 16. Interaction effects of fatigue and gender on Perceived Stability Ratings

Three-way interaction of CAP, fatigue and gender significantly affected peak velocity, sway area and peak-to-peak COP-COM AP (Table 8). It is evident that the peak velocity AP is larger for males and smaller for females post fatigue without the application of CAP (Figure 17; top). The peak velocity AP with the application of CAP, post-fatigue in males and females, both was more than peak velocity AP pre-fatigue (Figures 17; bottom). The difference between the two fatigue conditions, however, reduced with the application of CAP for males but increased in case of the females. Similar effects of three-way interaction of CAP, fatigue and gender were observed for dependent measures sway area and peak-to-peak COP-COM AP (Figures 18 & 19).

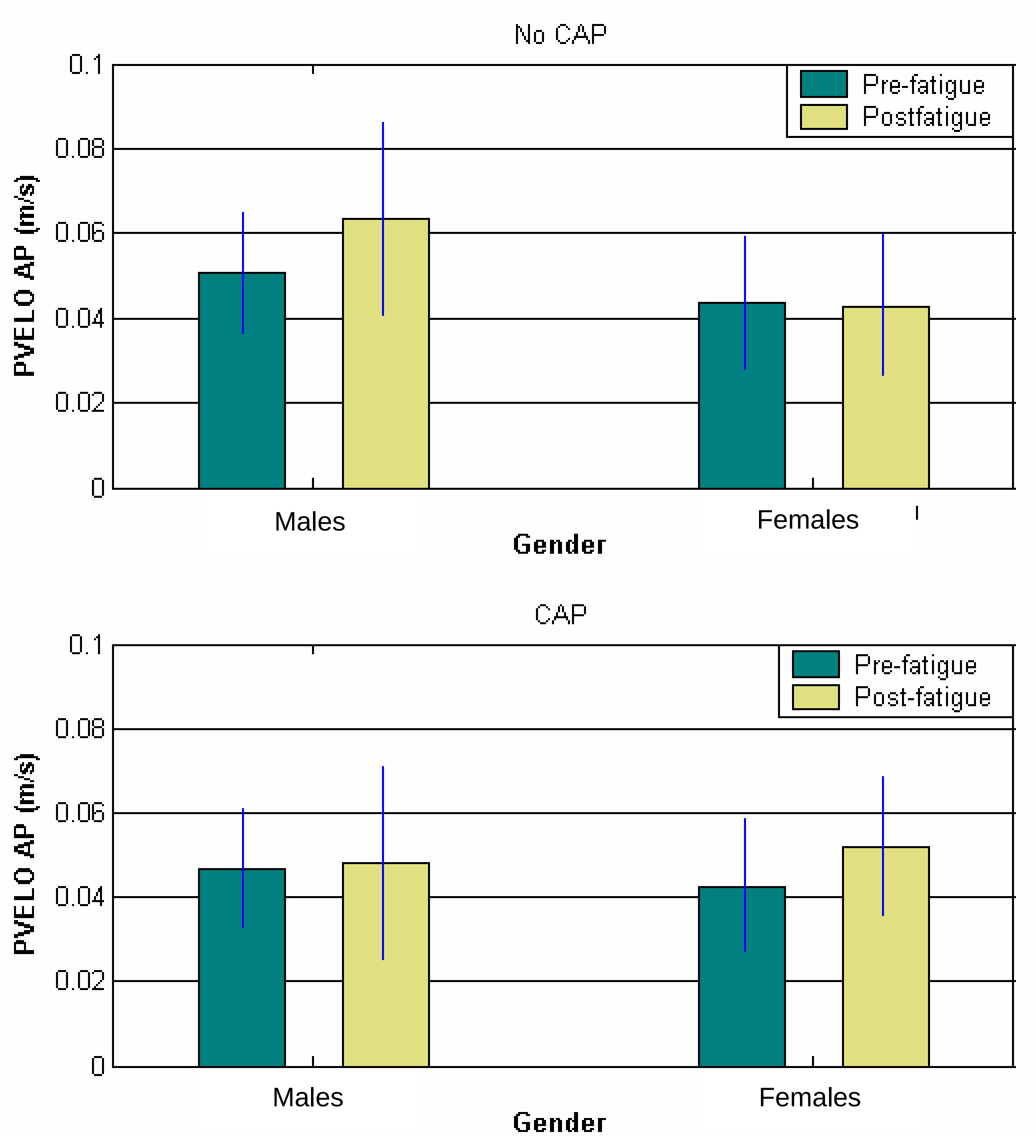

Figure 17. Three-way interaction between CAP, fatigue and gender on Peak velocity AP

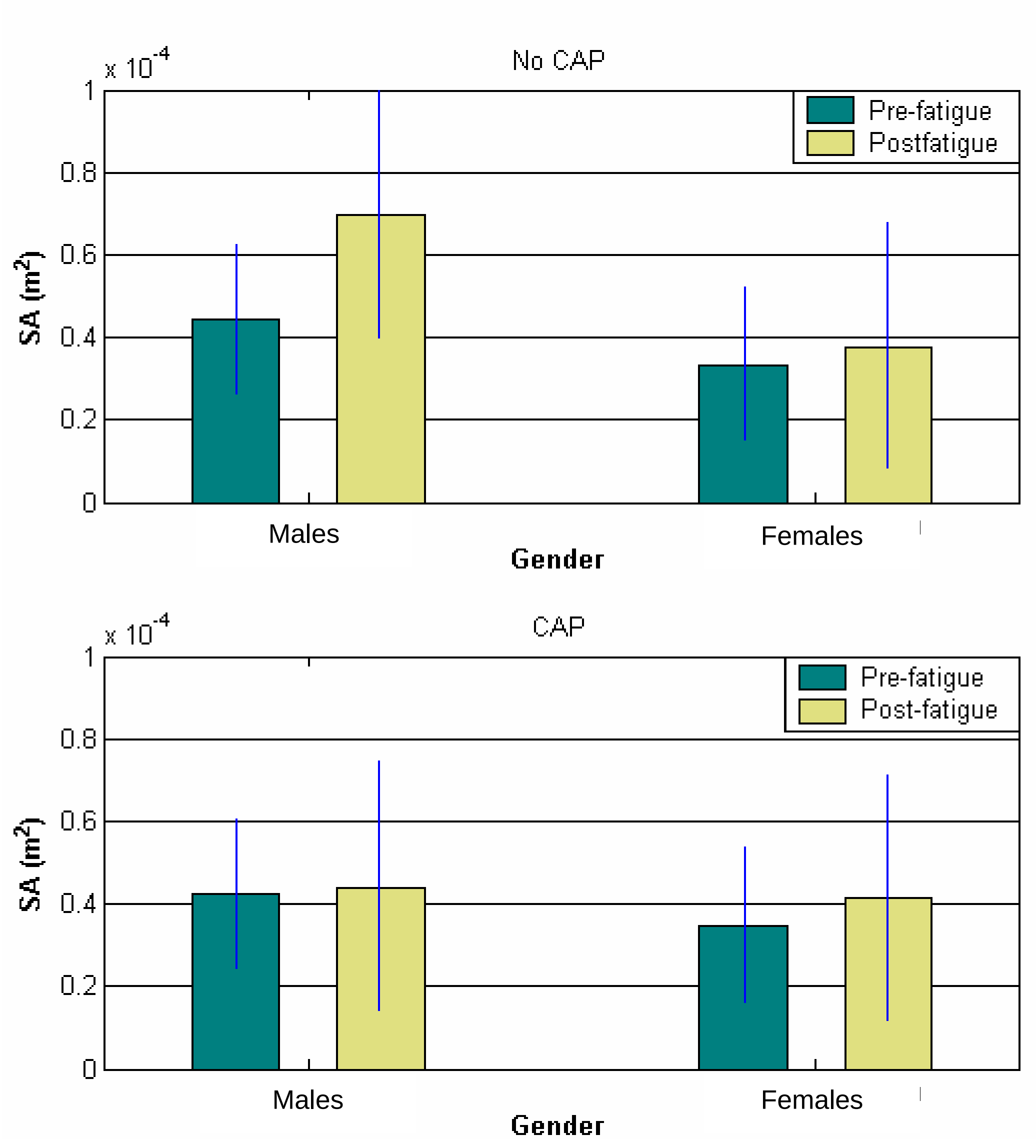

Figure 18. Three-way interaction between CAP, fatigue and gender affects Sway area

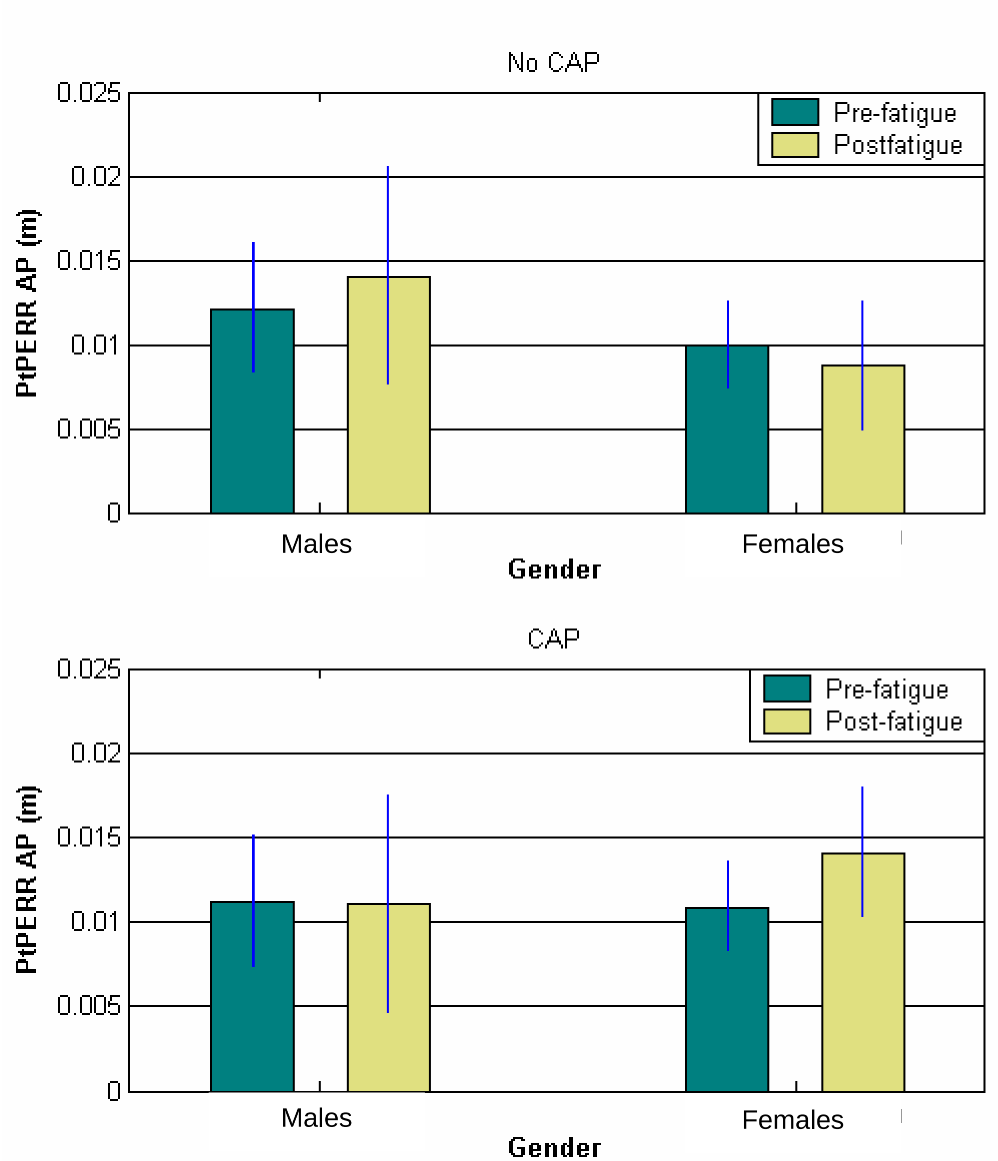

Figure 19. Three-way interaction between CAP, fatigue and gender affects Peak-to-Peak COP-COM AP

DISCUSSION

The purpose of the study was to investigate 1) whether the application of CAP improves postural control in individuals with lower PA, 2) whether fatigue adversely affects postural control in individuals with low PA and 3) whether CAP would be more effective in maintaining balance (measured indirectly using sway) when the postural control system is perturbed by inducing fatigue in individuals with lower PA.

Postural sway has been used previously to assess postural control. Sway can be assessed by using COP and/or COM based measures. One of the assumptions in analyzing sway using COP based measures is stationarity, though recently there have been reports demonstrating the non-stationarity of the COP signal. Ideally, a stationary time series should display consistent means and variances at least for the chosen sampling duration (Schumann, et al., 1995; Loughlin, et al., 2003; and Cao, et al., 2004). In other words, for the collected sample, the spectral characteristics of the signal would be time-invariant. Most measures used to investigate the postural sway characteristics (e.g. mean, RMS distance, RMS power, velocity of sway) assume stationarity. If the COP signal is in fact non-stationary, then the time-invariant methods used to describe sway characteristics could provide misleading results. Thus, to avoid any inconsistencies that might be associated with time-varying characteristics of the COP signal, DFA was used in the current study to analyze the COP signal in terms of non-stationarity. DFA is a method where a particular time series is first integrated, and then detrended, and finally an RMS of the integrated and detrended series is collected thereby removing the non-stationary behavior from the signal (Peng et al., 1995). Results from the study showed that for the most part the COP signal, when collected over a reliably large sample duration, behaves as a stationary time series. These results are in agreement with Duarte and Zatsiorsky (2000), involving large and reliable sample sizes, where the authors employed a similar fractal dimension approach based on the Hurst rescaled range analysis. Thus, the use of sway measures that assume COP as a stationary signal is likely valid to investigate postural control characteristics in this study.

The complexity involved in maintaining equilibrium (or balance) warrants the use of diverse sway measures in order to better understand the processes, subsystems and control strategies employed by the PCS. Balance is a concept used to describe a body in the state of equilibrium. In humans, this state of equilibrium occurs when the COM lies within the BOS (Pollock, et al., 2000). Since the human body can never be in a perfect state of equilibrium, sway measures have in the past been used to provide an insight on effectiveness of the state in which the current system exists. In order to validate sway as an indirect assessment for balance, investigations in the past have analyzed sway characteristics during a quiet stance in different scenarios. Overstall et al. (1977) demonstrated that elderly individuals with a history of falls exhibited increased sway when compared to individuals without any such occurrences in the same age-group. Another investigation reported a trend towards establishing postural sway as an indicator of a tendency to fall (Fernie, et al., 1982). The above-mentioned studies by Fernie et al., (1982) and Overstall et al., (1977) were able to associate sway prospectively and retrospectively with loss of balance (and eventually a fall incident). Sway has also been shown to increase in numerous conditions: in the absence of vision (Patla, 1997); while standing on a soft/compliant surface (Lord et al., 1993); upon inducing fatigue (Yaggie & McGregor, 2002) or upon introducing various perturbations, e.g. translation and/or rotation of the support surface (Allum et al., 1993).

Hence, sway measures collected during a quiet stance do appear to reflect the level and quality of activity performed by the PCS while maintaining balance.

The current study used a variety of sway measures to better understand some of the underlying behavior of the PCS. Correlation coefficients were determined for the 35 sway measures used in the study. Time-based measures were highly correlated with each other as well as RDIST (a distance-based measure). Velocity measures were inversely correlated with time-to-boundary measures which was expected (7). The subjective measure (PSR) was highly correlated with velocity measures. The latter suggests that in the absence of vision individuals perceive the speed at which they sway rather than the frequency or the distance of sway. The main findings of correlation analysis was that a combination of distance-based, time-domain, frequency domain, hybrid, kinematics based, combined COP and COM based and subjective measures should be used to determine the characteristics of the PCS. It should be noted, though, that the choice of dependent sway measures depends highly on the type of investigation.

There is an abundance of sway measures in the existing literature, but it is important to realize that each type of sway measure is associated with a particular aspect of postural and/or neuromuscular control. Distance-based sway measures have been used in the past to demonstrate the level to which the equilibrium is achieved, in other words distance-based sway measures reflect the efficacy of the PCS. Time-based sway measures on the other hand have been used to demonstrate the level of compensatory activity required to achieve a certain level of equilibrium (Maki et al., 1990; Prieto et al., 1996). Frequency-domain measures have primarily in the past been associated with identifying the discrepancies, if any, in the different sensory subsystems (namely the visual, vestibular and somatosensory) involved in maintaining equilibrium (Maki et al., 1990; Giacomini et al., 2004). The COP-COM signal, often termed the "error signal", has not only been used to account for the inconsistencies involved in separately considering the COP and the COM based measures but has also been used to assess the efficacy of the postural control system in achieving equilibrium (Winter, 1995; Benvenuti et al., 1999; Corriveau et al., 2001).

The significance of obtaining mean COP position and stiffness has mainly been towards evaluating the performance of the somatosensory system and how it might influence posture and orientation. A previous report has suggested that individuals tend to lean forward in situations when the lower extremity musculature is perturbed by fatigue (Madigan et al., submitted 2005). Various fractal dimension measures have also been considered in the past to describe either the feedforward and feedback loop model of the postural control system (Collins & DeLuca, 1993), or to define the long-range correlational process exhibited by the COP signal that could be an outcome of the human body being subjected to different physiological perturbations while attempting to obtain equilibrium (Duarte & Zatsiorsky, 2000). Kinematics based measures were obtained by calculating hip and ankle angles in the current study. The RMS values of these angles have been used to provide insight towards the postural strategy employed by individuals during quiet standing. Finally, the postural ratings are an individual's perception of the level of steadiness achieved (Schieppati et al., 1999). In addition to this, dividing the COP signal into uni-directional vectors in ML or AP has been shown to provide better understanding of postural control "hip" or "ankle" strategy, respectively, employed to maintain equilibrium. The use of a subset of various types of measures would provide a comprehensive knowledge about the performance of the PCS. Thus, the current study aimed to identify different aspects of PCS that might be affected by the application of CAP and/or the onset of fatigue.

A comprehensive investigation incorporating pressure application at a joint as an intervention towards mitigating the effects of localized muscular fatigue in maintaining balance has not been studied in detail; instead investigators in the past have analyzed various aspects of this issue separately. There have been numerous reports investigating the effects of pressure application in the form of taping, bracing and circumferential pressure on proprioception (MacGregor et al., 2005; Halseth et al., 2004; Batavia et al., 1999; Simoneau, et al., 1997; Perlau et al., 1995; etc.). The role or effects of fatigue/exercise on proprioception has also been a widely researched topic, especially over the past decade (Walsh, et al., 2004; Matre, et al., 2002; Hiemstra, et al., 2001; Ashton-Miller et al., 2001; Latanzio et al., 1997; Robbins et al., 1995; Sharpe & Miles, 1993; etc. to name a few).

Previous studies have shown that application of pressure might have differential affects on individual's ability to reposition joints (Perlau, et al., 1995; Callaghan, et al, 2002) as well as postural control (You et al., 2004). Thus, the current study subdivided participants equally into low and high PA groups based on the AAE obtained using a passive-active joint position sense test. This categorical classification of individuals into groups was based on the inherent accuracy/inaccuracy in achieving a preset angular position, and is assumed to represent proprioceptive deficits. However, this classification is by no means a diagnosis of participants' status.

The current study investigated the effects of application of CAP at the ankle joint, perturbed by localized muscular fatigue, on postural stability and control. There was a bias in the distribution of PA and gender among participants recruited in the study. Almost all males displayed higher PA while all females displayed lower PA. To address this clear confounding of the two variables, both PA and gender were used separately as the blocking variable. Most studies investigating the effects of gender on PA (Blasier et al., 1994; Taimela, et al., 1999; Djupsjobacka & Domkin, 2005) have reported non significant gender differences. However, Bjorklund, et al., (2000) investigated effects of fatigue and gender on PA of the shoulder joint and an active-active joint position sense protocol was used. From the review of findings presented here it is assumed here that PA categorization would have more substantial effects on postural control when compared to gender. In other words, results obtained with considering gender as a blocking variable likely were observed because of the bias that was inherent in the distribution, and thus results will be interpreted as due to differences in PA.

Results in the current study, involving the effects of localized muscular fatigue on postural stability were consistent with the existing literature. The study reported an overall increase in sway after fatigue was induced at the ankle joint, as illustrated by an increase in RDIST ML, a lateral shift of the absolute mean COP position in ML and lower PSRs post-fatigue. These findings are consistent with reports that have illustrated an increased postural sway upon inducing localized fatigue at the lower extremity (Lundin, 1993; Yaggie & McGregor, 2002; Gribble & Hertel, 2004). However, findings with respect to the effects of CAP on postural stability when considering individuals with low PA were contradictory to a few reports (You et al., 2004; etc.).

The main findings of the study were: 1) application of CAP had a detrimental effect on postural stability in individuals with lower PA, 2) Fatigue effects on postural stability in low PA group were more substantial as compared to high PA group, and finally, 3) application of CAP magnified the detrimental effects of fatigue on postural stability in individuals with lower PA, although its application improved the postural stability in individuals with high PA.

Lower extremity proprioceptive input is important in maintaining equilibrium and has been reported extensively in the literature. Lord et al. (1996) established proprioceptive deficits in the lower extremity as a risk factor for falls. Fitzpatrick & McCloskey (1994) reported that lower extremity proprioceptive signal affects sway characteristics, while Nardone & Schieppati (2004) reported that postural unsteadiness was associated with neuropathy in large and small-diameter afferent fibers. However, recently there have been reports which demonstrated that responses triggered after the application of a surface induced balance perturbation occurred simultaneously in gastrocnemius, trunk and neck muscles (e.g. Allum et al., 1993). If a response to a balance perturbation was in fact transmitted to different musculature simultaneously, this would challenge the original assumption that the human body behaves like an inverted pendulum and that the response to a certain perturbation travels from distal-to-proximal musculature via a number of links, which is also described as the so called "ankle strategy" (Nashner & McCollum, 1985). In another study it was reported that in situations when proprioception in lower extremity was selectively hampered, it did not trigger responses to accommodate for the discrepancy in proximal musculature (Allum et al., 1998). In other words, if the so-called "ankle strategy" were in fact used to transmit responses in distal-proximal fashion, it is likely that in lower PA group the lower extremity would trigger balance-correcting responses in the proximal musculature too.

From the review of concepts outlined in these reports, it is evident that, although lower extremity afferent proprioceptive information does play a role, this role is certainly not a necessary and sufficient condition for maintaining equilibrium.

Researchers have in the past compensated for the deficits of proprioceptive system by applying pressure in the form of taping, bracing and circumferential pressure to enhance the cutaneous and joint mechanoreceptors in the affected region (Halseth et al., 2004; Callaghan et al., 2002; Simoneau et al., 1997). The arguments supporting the use of these devices have been manifold. These devices: 1) provide a sense of mechanical stability to the joint sub-structure; 2) stimulate cutaneous and/or joint mechanoreceptors; 3) decrease pain in affected joint etc. The most valid argument for the current study appears to be stimulation of cutaneous and/or joint mechanoreceptors. The proprioceptive system in and around a joint sub-structure consists mainly of muscle, joint and cutaneous receptors. Recent research strongly favors the role of muscle receptors in the position and movement information, especially when the movement is in the mid-range of motion (Schmidt & Lee, 1999; Lephart et al., 1998). The sensory receptors located in central "bulging" region (or the muscle belly) are sensitive to the length of this region. This central region of the muscle contains gamma Ia fibers, which forms the basis of the neurological afferent pathway. The gamma Ia afferent fiber is not only sensitive to the length of the region, but also to the rate (velocity) at which this change in length takes place (Schmidt & Lee, 1999).

Previous reports investigating the effect of vibratory stimulation on postural sway (Andersson et al., 1998) and body orientation (Kavounoudias et al., 1999) have reported that the during vibratory stimulation the musculature of interest is lengthened thereby affecting sway or body orientation. The application of mechanical pressure has a similar effect; it compresses the musculature, thereby lengthening the muscle spindles involved. This lengthening of the muscle spindle perturbs the otherwise constant stimulus muscle spindle structure, and readjusts the gain of the musculature (Ashton-Miller et al., 2001), thereby rendering the musculature more sensitive. The readjusted gain now defines the new firing rate of the muscle spindles as well as the basis for the sensory afferent information, until the pressure is removed or in presence of a perturbation.

As has been mentioned previously, proprioception can be broadly categorized into JPS and kinesthesia. There have been numerous reports investigating which test/method should be used to measure proprioception and/or kinesthesia. Grob et al. (2002) defined proprioception as a sum of kinesthesia and JPS and reported low correlations between 5 different joint proprioception assessment paradigms, suggesting different tests designed to measure proprioception might just measure certain aspects of it. The same study also argued that joint position tests may be more objective, whereas joint motion sense tests might be more reliable, in measuring proprioception. There is still a lack of information suggesting which test/s of proprioception might be more objective or valid when investigating the effects of proprioception on balance. You et al. (2004) reported that CAP reduced sway (improved balance) in ML directions in individuals with lower PA. The investigators in the study used an active-active paradigm to measure joint position sense, but collected only 10s of postural sway data. Authors in the same study also found that the sway in AP direction did increase after the application of CAP, however, this increase was not significant. On the contrary, a recent report investigating the affects of CAP on JMS concluded that CAP impaired JMS (Pline et al., 2005). Due to the complex nature of the hypothesis proposed, it was difficult to choose a PA protocol. It has been shown previously that JPS is not only more sensitive in identifying the effects of fatigue (Lephart et al., 1998) but is more objective measure of proprioception when compared to JMS protocols (Grob et al., 2002). However, the JMS protocol is a more reliable measure of proprioception (Grob et al., 2002) and could be more effective in categorizing individuals when postural control is analyzed.

Also of note here is that tests to evaluate proprioception measure the performance of the afferent pathway while postural control measured via sway is an indicator of the performance of both the afferent and efferent pathways (Lephart et al., 1998). From the trends reported in the current study, it is evident that individuals with low PA performed similar or better when comparing the results from the COP based measures to individuals with high PA (had similar postural sway). Thus, it could be concluded that the individuals in low PA group have somehow compensated for or acclimatized to the impairment so that the efferent pathway (measured via the postural control) is not adversely affected.

A passive-active ankle joint position reproduction error assessment was performed in the current study. The assumption was that inability to successfully reposition the joint sub-structure would affect sway characteristics. The use of passive protocol in determining JPS eliminates inputs from various other receptors such as Ruffini endings and Golgi tendon organs (Lephart et al., 1998). If muscle receptors are in fact the only receptors that contribute to the sense of position and motion of a joint (Schmidt & Lee, 1999) as is commonly believed, it is beneficial to use the passive positioning protocol. However, more recent information states that use of muscle receptors as position sensors has potential disadvantages. The muscle receptors associated with the joint substructure providing length (or position and velocity) information are ambiguous in terms of the source of the information. In other words, it is difficult to decipher whether the information is exteroceptive or proprioceptive (Proske, 2005). If this were true, then use of passive positioning protocol for JPS test is not advised. Until more concrete information is obtained on whether position or velocity information is provided by muscle or joint receptors or a combination of both, it is difficult to say which protocol (active or passive) is more valid in detecting whether or not deficits in proprioception at a joint will adversely affect postural control.

Hypothesis 1

Both PA and gender considered as blocking variables separately provided similar results with respect to CAP and PA interactions. Results from the current study on the effects of CAP on postural sway contradict previous findings in the literature. Individuals with low PA swayed more after the application of CAP as compared to high PA group. However previous studies have shown that application of tape in individuals with lower PA enhances proprioception. Possible reasons for this deviation could be: 1) JPS might be an objective protocol to measure deficits in proprioception, but it is difficult to say whether it is a valid measure of proprioceptive deficits in studies investigating the effects of proprioceptive acuity and pressure application on postural sway. 2) Allum et al. (1993) reported that responses triggered by inducing balance perturbations occurred simultaneously in gastrocnemius, trunk and neck muscles thereby challenging the “ankle strategy” hypothesis. If the responses to balance perturbations are in fact transmitted simultaneously to the entire musculature, then in the case of application of CAP would provide better stability to the ankle joint, but not to the proximal musculature. This could be a reason for more sway in individuals with lower PA after CAP, since upon application of CAP stimulus would only be enhanced in the distal musculature, whereas the proximal musculature would not be adjusted or modified. 3) Due to the lack of evidence on the type of proprioceptive tests valid for sway studies, it is recommended that multiple measures of proprioceptive acuity should be used as different tests might reveal different aspects of proprioception. This is also supported by the evidence that different measures of proprioception are poorly correlated with each other (Grob et al., 2002). 4) You et al. (2004) reported CAP as an effective tool in reducing sway in individuals belonging to the low PA group proprioception, but they used a very small sample, just 10 s, to analyze sway. With a reliably large sample size the current study reports that sway increased in individuals in lower PA group. 5) It is also possible that in order to compensate for proprioceptive deficits in the dominant leg low PA group placed more emphasis on the non-dominant leg. 6) As this was an exploratory study, proper estimation of sample size could not be obtained. For the same reason, COP based measures in frequency domain, kinematics based measures like RMS Ankle angle in ML and RMS Hip angle in AP demonstrated trends in support of the above-mentioned hypothesis, but these trends were not significant. Use of these measures is recommended in future studies where investigators are interested in investigating the effects of proprioceptive deficits (assessed as lower PA) on postural control and sway. Additionally, the use of frequency-domain COP based measures could be supported by the fact that previous studies have reported different frequency bandwidths to be sensitive to the different afferent input pathways (visual, vestibular, proprioceptive) leading to equilibrium maintained by the postural control system (Nagy et al., 2004; Oppenheim et al., 1999; Golomer et al., 1997). 7) MacGregor et al. (2005) found change in vastus medialis obliquus (VMO) surface amplitude varied significantly with directional application of pressure at the knee. This would imply that a greater control of pressure could be needed to properly enhance proprioception. If the PA is lower for plantar-flexion joint position sense then pressure should likely just be applied in AP direction, thereby sensitizing and stimulating only the relevant muscle spindle system.

Hypothesis 2

None of the dependent measures illustrated any significant interaction effects of fatigue and PA when PA was used as a blocking variable. However when gender was considered as the blocking variable median power frequency in ML, mean COP position in ML and postural stability ratings were found to be significantly affected by fatigue and PA interactions. To reiterate, in light of previous findings the results from considering gender as a blocking variable likely existed because of the underlying bias and is discussed accordingly.

Previous studies have reported that localized muscle fatigue induced at lower extremity does affect postural sway and thereby compromises the postural control system (Yaggie & McGregor, 2002; Lundin et al. 1993; etc.). Localized muscle fatigue induced at a more proximal musculature like the hip and torso has also been shown to affect postural sway (Gribble & Hertel, 2004; Davidson et al., 2004). In addition to affecting sway, lumbar extensor fatigue has also been shown to have detrimental affects on kinesthesia (Pline et al., 2005; Taimela et al., 1999) as well as joint position sense (Bjorklund et al., 2000).

The above-mentioned studies and various other findings provide sufficient evidence that introduction of localized muscular fatigue does impair proprioception and neuromuscular control, thereby impairing the afferent information as well as the efferent responses. It is difficult to suggest a definitive explanation as to how muscle fatigue affects postural control. Possible explanations include: 1) muscle fatigue occurs due to the loss of K^+ ions which causes a decrease in muscle spindle excitability; 2) muscle fatigue leads to an increase in the threshold for muscle spindle discharge which causes a change in co-activation of the alpha and gamma motor neurons as well as a desensitization of muscle, joint and cutaneous receptors (proprioceptive system); 3) fatigue causes an increased latency in motor unit firing, which eventually impairs neuromuscular control (Enoka & Stuart, 1992).

One of the major arguments supporting the adverse effect of ankle fatigue on postural sway is the apparent loss of proprioception in the musculature caused by inducing fatigue. Another possible argument comes from the hypothesis commonly known as the "ankle strategy" (Nashner & McCollum, 1985) which states that the compensatory responses to a balance perturbation are transmitted in a distal-proximal fashion from the ankle upwards one link at a time. If such a strategy does exist, then it is likely that a fatigue perturbation at the ankle joint would either affect the response or delay it; thereby causing a readjustment in the input stimulus either at the spinal cord or at the CNS and thus postural control would be adversely affected.

Results from the current study show that, pre-fatigue, MDPF AP was higher for the low PA group than the high PA group, and that it further increased for individuals with lower PA post-fatigue. Thus, the lower PA group swayed more frequently than the higher PA group. This is in accordance with the hypothesis that effects of fatigue would be more severe in low PA group. RDIST ML however, was found to decrease in low PA group pre- as well as post-fatigue when compared with higher PA group. This is counterintuitive as with deficits in proprioception the level of steadiness achieved should be less than that obtained without such deficits. However, it is possible that individuals with lower PA just incorporated other sensory pathways to compensate for the deficits in proprioception shown by the JPS test.

Mean COP position in ML was found to move further away from origin, pre-fatigue, in females as compared to males. This could either be due to the fact that individuals with lower PA adopted a stance that would compensate for the deficits in lower extremity proprioception, or could be due to the fact that reduced stiffness in ML in females as compared to males (a trend). Males (high PA group here) on average provided better PSRs than females (low PA group here), pre-fatigue. This was an expected outcome, however, the PSRs improved for females post-fatigue. PSR in individuals is influenced by the cognitive processing required by the CNS to maintain postural stability or steadiness (Schieppati et al., 1999), and it could be that for individuals with low PA group the cognitive processing required was elevated, making it difficult for them to determine a proper PSR post-fatigue.

Hypothesis 3

The PCS comprises the set of activities performed by the CNS to integrate inputs from the visual, vestibular and somatosensory systems. This study eliminated visual inputs by incorporating the eyes closed condition throughout the course of the experiment. Since the vestibular system was not hindered, it is assumed that vestibular inputs to the CNS were unchanged. Thus, the study was aimed at assessing the effects of modifying the somatosensory system (first by inducing fatigue to the lower extremity musculature and then by applying CAP to stimulate and sensitize the proprioceptive inputs) on the postural control system. In other words, the current study aimed at investigating whether modifying the sensory stimulus affected the motor response. Additionally it was ensured that individuals were categorized according to the performance of the afferent or sensory pathway. Information from the different input subsystems are processed at 3 different levels, namely 1) spinal cord, 2) brain stem and finally, 3) higher brain centers. The spinal cord is responsible for stabilization of the musculature as well as synergizing muscle activation. The brain stem integrates the 3 subsystems to maintain balance and posture, while, the higher brain centers are associated with the ensuing action and/or motion. In this study the application of CAP post-fatigue in individuals belonging to the higher PA group resulted in decreased peak velocity, and decreased sway area, when compared to the no/without CAP condition. This likely would imply that application of CAP did in fact sensitize the somatosensory and proprioceptive inputs to the spinal cord and finally to the CNS thereby showing improvement in the motor mechanism as indicated by decreased sway (or better postural control). However the effects of application of CAP were entirely contrary in individuals with lower PA. In low PA group, application of CAP actually increased sway post-fatigue. This could likely mean that although CAP sensitized the somatosensory inputs to the spinal cord, it either: 1) did not mitigate the effects of fatigue; 2) CAP actually sensitized an entirely different aspect of the somatosensory input and did not have any effects on the muscle activation patterns already hindered by fatigue; 3) It is likely that since fatigue affects both the somatosensory inputs and efferent motor performance CAP was able to compensate for the somatosensory deficits or impairements but had no influence on the post-fatigue motor performance. Extending this argument it could be that low PA group somehow were more substantially affected by fatigue in terms of the efferent motor mechanism and thus the increase in sway post-fatigue with CAP; 4) in individuals with lower PA there exists a greater likelihood of overcompensation. In other words, the application of CAP on ankle musculature already perturbed by fatigue in low PA group somehow overcompensated for the proprioceptive deficits thereby resulting in heightened motor responses and larger sway; or it is also possible that 5) Individuals with lower PA have somehow compensated for this deficiency in such a way that they do not rely on the afferent proprioceptive signals while maintaining balance (perhaps relying more on vestibular pathways). The support for this argument is provided by the fact that, individuals with lower PA showed almost no change in velocity of sway (and sway area) post-fatigue without CAP, whereas, high PA group had increased peak sway velocity (and sway area) for the same condition. On the other hand, the peak velocity of sway (and sway area) remained the same for high PA group with CAP whereas the low PA group had higher peak sway velocity (and sway area) for that condition (Figures 17 & 18). Finally, the median power frequency in AP did increase for individuals with low PA group post-fatigue indicating that the individuals with low PA group did sway with higher frequency components post-fatigue (Figure 13). As has been previously mentioned the frequency spectrum analysis of COP signals provides an assessment on the performance of various afferent pathways involved in postural control. It is possible that although time-based (velocity and mod. time-to-boundary) and distance-based (RMS distance) measures were ineffective in identifying the lack of (or weaker) proprioceptive information in maintaining postural control in individuals with lower PA, the frequency-based measure (Median power frequency) was able to identify this discrepancy in the postural control system.

Finally, previous studies have demonstrated that increased sway velocity and larger sway area (Fernie et al., 1982) may be a predictor of risk of falling. Fallers tend to exhibit higher values for RMS distance in ML and AP (Kirshen et al., 1984; Maki & McIlroy, 1996). Additionally, individuals suffering from neuromuscular disorders like Parkinson's disease have been shown to have larger sway area and lower values of mean time-to-boundary (van Wegen et al., 2001; Hagiwara et al., 2004). Participants in the current study did illustrate a considerable increase in mean sway velocity and sway area post-fatigue, indicating that fatigue could be an occupational risk factor for falls. Furthermore, individuals with higher PA indicated a decrease in sway area and peak sway velocity in AP, post-fatigue with CAP when compared to without CAP condition.

There were several limitations in the study. Fatigue at the ankle was induced only on the dominant side, due to equipment limitations, and CAP was also applied to the dominant side but the stance acquired for quiet standing was bilateral. It is possible that bilateral fatigue induced at the ankle might show more substantial changes in post-fatigue sway measures, and same could be true for pressure application too. Several reasons motivated the choice of bilateral stance, including occupational relevance and the fact that most individuals can only maintain unilateral stance for brief periods. It is possible that individuals in the post-fatigue trials placed more emphasis on the non-dominant leg while standing, thereby mitigating the effects of fatigue. The possibility of increased emphasis on non-dominant leg during a bilateral stance is further accentuated by a considerable (approx. 46%) greater lateral shift in mean COP position post-fatigue. Also the study did not investigate the affect of plantar pressures during bilateral stance which might provide additional insight on postural sway patterns (Gatev et al., 2001). This being an exploratory study, the power approach or effect size estimation could not be performed in order to calculate the adequate number of participants needed to investigate the important interaction effects, and a higher alpha level of $p < 0.1$ was chosen to enhance power. For future work, this study could serve as a basis for investigations involving combined effects of pressure application on different joints, deficits in proprioception, and localized muscle fatigue on postural sway.

RECOMMENDATIONS

The current study presents a few recommendations for future studies investigating the effect of proprioception, and localized muscle fatigue on postural sway. It is recommended that 1) before incorporating sway measures as an indirect assessment of balance, the COP signal be investigated for the property of stationarity; 2) multiple assessments of proprioception and kinesthesia be incorporated while assessing PA; 3) while bilateral induction of fatigue and bilateral use of CAP should be considered; and finally, 4) in addition to sway measures, plantar sensitivity or pressures of the foot should also be investigated during quiet stance.

CONCLUSION

The study explored the effects of CAP on postural sway. The major findings of the study were that the application of CAP deleteriously affected postural sway in individuals with low PA, fatigue effects on postural sway were more substantial in individuals with low PA and finally, instead of mitigating the effects of fatigue application of CAP actually magnified the effects of fatigue in individuals with low PA as demonstrated by larger postural sway. CAP was found to be ineffective in mitigating the effects of fatigue on postural sway in low PA group, however it should be noted that PA was determined using a single protocol whereas joint proprioception is a combination of various activities from various receptors. Application of CAP, however, did improve postural sway in high PA group post-fatigue. CAP was also helpful in reducing postural sway in individuals with high PA. The results from the study could be used in evaluating CAP and other strategies towards control of fatigue-related falls. Results in the study could also be helpful in designing larger and more comprehensive experiments investigating ergonomic interventions towards fatigue-related falls, with a focus on determining PA using motion sense or a combination of different position and/or motion sense tests and bilaterally inducing ankle fatigue and applying CAP.

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APPENDICES

APPENDIX A: Forms for screening

APPENDIX B: Informed Consent form

APPENDIX A

PARTICIPANT INFORMATION AND SCREENING FORM

Participant #: _________ (filled out by the experimenter) **Date:** ____________

Part I

Name : __________________________ (Last), _____________________ (First)

Phone : _______________ (H) ________________ (O) Email: _______________________

Address : ____

Birth date (mm/dd/yy): __________________ Age : ___________ years old

Gender (please circle) : Male - Female

Ethnic category: ☐ Caucasian ☐ African American
☐ Asian ☐ American Indian/Alaska Native
☐ Other ☐ Native Hawaiian/Pacific Islander

Racial category: ☐ Hispanic/Latino ☐ Not Hispanic/Latino

Occupation : ________________________ (Current) ________________________ (Previous)

Length of time at present occupation : ____________ years

Part II

Dominant arm : ☐ Right ☐ Left

Dominant leg : ☐ Right ☐ Left
(ask experimenter if not sure)

Frequency of physical exercise of 15 min or more: ________ days/week

Please describe types of exercise : ____

Are both of your legs the same length: ☐ Yes ☐ No

Is your corrected vision 20/20? ☐ Yes ☐ No

Do you currently use any sedatives, tranquilizers, or muscle relaxants? ☐ Yes ☐ No

Please list any current medical conditions and/or use of medications:

Part III

Have you had any history of the following (Please circle 'Y' or 'N')?

1. Y – N : Shoulder and upper extremity problems during the past year
2. Y – N : Upper/lower back problems during the past year
3. Y – N : Falls in the past three years
4. Y – N : Lower extremity injuries in the past three years
5. Y – N : Joint replacement or joint fusion.
6. Y – N : Being treated with corticosteroids for any condition
7. Y – N : Problems caused by arthritis, muscle problems, or broken bones, etc. that limit your ability to walk or bend your joints
8. Y – N : Ear infection or drainage from the ear in the past 6 months
9. Y – N : Severe head injury, concussion, or been 'knocked out'.
10. Y – N : Problems with coordination, dizziness, or loss of balance (in the past 12 months) that seemed to occur frequently or lasted for an extended period of time.
11. Y – N : Neuromuscular/neuromotor problems
12. Y – N : Any other disorders, illnesses or injuries that you feel might interfere with this study.

For any 'yes' answer above, please describe the time, type, extent, duration, and limitations on your daily activities.

Part IV

Availability (including evening hours)

Monday: ______________________________________

Tuesday: ______________________________________

Wednesday: ______________________________________

Thursday: ______________________________________

Friday: ______________________________________

Weekends: ______________________________________

APPENDIX B

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants
In Research Projects Involving Human Subjects

Title of the Research Study
"Determining the relationships between different exercises and balance during static standing"

Investigators
Maury A. Nussbaum, Ph.D. 231-6053 – Department of Industrial and Systems Engineering
Michael L. Madigan, Ph.D. 231-1215 – Department of Engineering Science and Mechanics

I. Purpose of this Study
The purpose of this research study is to investigate the relationships between different exercises and balance during static standing. Falls from heights are a major problem in both industry and general society when measured in terms of human suffering and economic losses. Work tasks performed by various muscle groups may influence standing balance differently, and joints of the ankle, knee, lower torso, and shoulder are of particular interest due to their possible contributions to standing balance. Findings from this research study will provide a better understanding of the mechanisms associated with standing balance, and contribute to the development of practical interventions aimed at decreasing the risk of falls.

II. Procedures
A total of 40 adult participants will be used for the study.

The study will take place in either the Industrial Ergonomics and Biomechanics Lab (Department of Industrial and Systems Engineering) or the Musculoskeletal Biomechanics Lab (Department of Engineering Science and Mechanics). Upon arriving, you will be briefed of the study protocol, asked if you have any further questions, and asked to sign this informed consent form.

Prior to the experiment, several non-invasive position sensors will be placed on your body using double-sided tape. For this purpose, you will be required to wear a special clothing. The experimental session will be videotaped to help the investigators analyze your movement patterns during the experiment.

At the start of the experiment, you will be asked to stand still on a force platform for 75 seconds. You will then perform an exercise on a Biodex System (similar to a health club-type exercise apparatus) to work out muscles of your ankle, knee, lower torso, or shoulder. Immediately following the exercise, you will return to the force platform for a second balance measurement. At this moment, balance measurements will be performed identically to the initial measurement procedure. This procedure will also be repeated every 5 minutes for 30 minutes to assess the recovery of balance following the exercise. The experiment is expected to take approximately 1.5 - 2 hours to complete.

In addition to an initial 1-hr practice session (which will also be used to obtain baseline measurements), you will be asked to complete the experiment in four separate occasions, once for each fatigue location (ankle, knee, lower torso, and shoulder). Each of these will be on different days separated by at least 48 hours.

III. Risks

The risks involved in this study are minimal. The overall physical exertion required during this experiment is not significantly larger than that required during common manual labor. You may feel residual muscle soreness, which will typically be gone within a few hours or within a day. If you experience any substantial amount of pain following an exercise, you should contact us immediately.

IV. Benefits

You will receive no direct benefit from participating in this study. The scientific community will benefit through the additional information that is expected to result from the completion of this study. This information will contribute to fall-related biomechanical knowledge that will be used to develop intervention techniques to prevent falls from heights.

No promise or guarantee of benefits has been made to encourage you to participate.

V. Extent of Anonymity and Confidentiality

The results of this research study may be presented at meetings or in publications. Your identity will not be disclosed in those presentations. All participants will be identified based only on their unique identifying number. Only the investigators and experimenters involved in the research will have access to these identifying numbers. The video recordings from this study will be analyzed and stored in the labs under the supervision of the investigators. Some photographs or video recordings may be shown to other scientists at the University or at scientific conferences.

VI. Compensation

You will be paid $10/hour for your participation in this study. A bonus in the amount of $20.00 will be provided at the completion of this study.

VII. Freedom to Withdraw

Your participation in this research study is voluntary. Refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled. You are free to withdraw from the study at any time without penalty.

VIII. Approval of Research

This research project has been approved, as required, by the Institutional Review Board for Research Involving Human Subjects at Virginia Polytechnic Institute and State University.

IRB Approval Date: 01/15/2004

Approval Expiration Date: 01/14/2006

IX. Participant Responsibilities

I voluntarily agree to participate in this study and to follow the responsibilities listed below:

a. To inform the investigator/experimenter as early as possible about a desire to discontinue participation in the study.

b. To inform the investigator of any medical conditions that might be adversely affected by the experiment, or those that might interfere with results of the experiment.

X. Participant's Permission

I have read and understand the Informed Consent and conditions of this project. I have had all my questions answered. I hereby acknowledge the above and give my voluntary consent:

______________________________	______________________________	________________
Signature	Printed name	Date
______________________________	______________________________	________________
Witness/experimenter	Printed name	Date

Should I have any pertinent questions about this research or its conduct, and research participants' rights, and whom to contact in the event of a research related injury to the participant, I may contact:

Principal Investigator:	Maury Nussbaum, PhD	231-6053	nussbaum@vt.edu
Co-Investigator:	Michael Madigan, PhD	231-1215	mlmadigan@vt.edu
Student Researcher:	Navrag Singh		nbs@vt.edu
Chair, IRB:	David M. Moore, DVM	231-4991	moored@vt.edu

This Informed Consent is valid from 01/15/2004 to 01/14/2006.

VITA

Navrag Singh completed his Master of Science degree in Mechanical Engineering from University of Missouri-Rolla in 2002. He joined the Master of Science program at Virginia Tech in Industrial & Systems Engineering with a focus in Human Factors, Ergonomics and Biomechanics. While completing his Masters, Navrag worked as a research assistant in Industrial Ergonomics and Biomechanics laboratory. He was an active member of the VT HFES student chapter and was instrumental in organizing sports activities for the chapter. He was the organizer and captain of the volleyball and softball chapter teams which participated in intramural sports competition on campus. He was also a member of Alpha Pi Mu.