Effects of Localized Muscle Fatigue on Risks of Occupational Slips and Falls

Sponsor
National Institute for Occupational Safety and Health (NIOSH)

Grant #
R01 OH009222

Performance Period
September 01, 2009 – August 31, 2014*
*includes a 1-year no-cost extension

Final Report

by

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November 11, 2014

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ABBREVIATIONS

NORA – National Occupational Research Agenda

RCOF – Required Coefficient of Friction

LBD – Lower Back Disorder

LMF – Localized Muscle Fatigue

MVE – Maximum Voluntary Exertion

LANL – Los Alamos National Laboratory

F_h *– Horizontal ground reaction force*

F_v*- Vertical ground reaction force*

DAFW - the number of days away from work

BLS – Bureau of Labor Statistics

SOII - the Survey of Occupational Injuries and Illness

OIICS - Occupational Injury and Illness Classification System

COM - Center-of-Mass (COM)

IMU – Inertial Measurement Units

SVM – Support Vector Machines

PoV - Proportion of Variance

ROC - Receiver Operating Characteristic curve

RBF - Radial Basis Function

CNS – Central Nervous System

HCV-heel contact velocity,

WV- walking velocity

TA- whole body COM transitional acceleration

$Kneemom_{peak}$*- knee moment peak*

ABSTRACT

Injuries associated with slip and fall accidents continue to pose a significant burden to industry, both in terms of human suffering and economic losses. A majority of occupational falls leading to injuries and deaths are a result of foot slippage and are typically experienced by laborers. The literature provides convincing arguments that localized muscle fatigue can disrupt the quality of the signal from the periphery for effective balance control during slip perturbations, and increase the risk of slips and falls. In this study, localized muscle fatigue and its effect on gait and balance leading to increased fall risks were investigated. This information can provide a better understanding of the mechanisms involved in fall recovery and slip initiating characteristics that may influence prevention strategies associated with occupational falls.

Two laboratory studies were conducted to quantify the effects of localized muscle fatigue on slip propensity and balance recovery. The first experiment evaluated the effects of localized muscle fatigue on the slip initiation process while walking over a non-slippery surface. Distal limb muscles (ankle plantarflexors, knee extensors) and a combination of ankle, knee, and hip muscles were fatigued independently on different weeks. Additionally, the effects of floor inclination, load carriage, work pace (i.e., walking speed) and age was ascertained. The second experiment evaluated the effects of localized muscle fatigue on the balance recovery process following slips induced by walking on an unexpectedly slippery surface. Distal limb muscles as well as proximal muscles (i.e., low back) were fatigued independently on different weeks. The effects of floor inclination, work pace, and age was ascertained. The work addresses several NORA Priority Areas in the context of work-related traumatic falls. The main Priority Areas are: 1) Risk Assessment Methods; 2) Control Technology; and 3) Intervention Effectiveness Research.

In this final report, the specific aims concerning the “slip-initiation” characteristics as well as the “slip/fall-recovery” characteristics will be discussed in lieu of gait changes associated with lower leg localized muscle fatigue and, slip propensity changes associated with two different types of lower extremity muscle fatigue conditions (i.e., squat and leg extensions on the Biodex). Additionally, most recent results regarding the ankle fatigue condition are introduced. Future researches concerning the slip initiation characteristics are further elaborated. Afterwards, the effects of fatigue on slip recovery characteristics are further explored. Here, two studies are discussed in lieu of fall recovery responses to a slip perturbation associated with quadriceps fatigue and back muscle fatigue conditions.

SIGNIFICANT (KEY) FINDINGS

In the current study, we aim to monitor kinematics of walking in unconstrained environments. We hypothesize that lower extremity muscle fatigue will influence walking behavior and this subtle changes in gait can be classified by supervised machine learning techniques such as support vector machines. The results indicated that lower extremity muscle fatigue condition influenced gait and loading responses (i.e., jerk cost). Overall, the fatigue effects were observed on various gait parameters. Specifically, heel contact velocity and walking velocity was faster when fatigued with longer step length. Additionally, single support time was reduced with decreased toe clearance height. It appears that when fatigued, we may revert to a safer gait strategy – i.e., decreasing single support time to increase the double support time for better dynamic balance maintenance. In order to do so, individual may needed to walk faster with higher heel contact velocity. Although RCOF was not significantly affected by fatiguing, given a slippery floor surfaces, slip/fall risk may increase. Additionally, results suggest that single stance duration is decreased in post-fatigue walking trials. During stance phase of the gait cycle, proprioceptive input from extensor muscles and mechanoreceptors in the sole of the foot provide load information [11] to the central nervous system. Fatiguing of the muscles around a joint inhibits the joint's neuromuscular feedback and synergism between joint proprioception leading to instability and gait changes [15-23]. Thus, the reduced stance duration decreases foot-loading information through afferent sensory and proprioceptive mechanoreceptors, such as Golgi-tendon units, muscle spindles, and joint receptors, and may have adversely influenced motor control of the lower extremity during walking. As such, results indicate that fatigue adversely influences gait and this change. We also found that SVM classifier incorporating trunk kinematic signals during walking has an excellent potential to predict fatigue status intra-individually (~97% accurate predictions) as well as inter-individually (~ 90% accurate predictions).

Furthermore, in the slip/fall recovery experiments, bilateral fatigue was induced by performing repetitive isokinetic knee extension using the quadriceps. The findings from this study indicate that localized muscle fatigue is a potential risk factor causing slip-induced falls.

Results also indicated that, in comparison with values during normal walking, lumbar kinematics, lumbosacral kinetics, lumbar muscle activations, and lumbosacral reaction forces were all substantially increased during a slip event. Observed levels of muscle activity and lumbosacral reaction forces suggest the potential for low back injury during a slip event. Outcomes of this work may facilitate the identification and control of specific mechanisms involved with low back disorders consequent to a slip.

The present results thus suggest that, in additional to causes subsequent to a slip (e.g., contact with the ground from a fall), low-back injury could result from high levels of forces generated during the slip event itself. Outcomes of this study may help to identify potential slip-related mechanisms in the development of LBDs as well as future preventive approaches.

In summary, LMF of the quadriceps affects various kinematic and kinetic gait variables that are linked with a higher risk of slip-induced falls and, therefore, can be considered as a potential risk factor for slip-induced falls. Additionally, the results also indicate that LMF of the knee extensors caused a delayed response in producing joint moment and increasing the base of support using the trailing limb. One of the limitations of the study was that each participant reached their fatigue level at a different time. These limitations can affect the results due to the difference in the fatigue level of each individual. Additionally, quadriceps musculature consists of four different muscle groups (rectus femoris, Vastus lateralis, Vastus intermedius and rectus femoris). The current study did not isolate muscle fatigue to each of the muscles but as the quadriceps muscle group as a whole. There might be limitations as rectus femoris muscle does not fatigue in the same way as other muscles in this group. Although implicated, the 60% of baseline MVE as a fatigue state prior to testing ensured that all participants were fatigued at similar levels.

Future research will investigate the effects of LMF of multi-joint fatigue (i.e. hamstrings, ankle plantar flexors) on slip events in a real-world job scenario. However, results from the present study can be used as preliminary information on the specific gait and slip parameters that are sensitive to LMF. Other potential areas for further research include evaluating the effects of rest breaks and recovery and effects of age on fatigue.

TRANSLATIONS OF FINDINGS

Applying this knowledge, a novel slip simulator training program was developed to enhance individuals' ability to recognize and take appropriate actions to various fall hazards at work places utilizing the "kinetic learning" principle. The results indicated a beneficial effect of slip simulator training in reducing fall accidents. Currently, several more companies including the UPS, DOE, Diageo, Snap-On Tools, Sandia National Laboratory, Consumer Energy, GE, BP and Los Alamos National Security and others have adopted the training program.

OTUCOMES / IMPACT

Impact of Slip Simulator Training at LANL

Slip Simulator training started in February 2011 for students in the Hazardous Waste Operations training class utilizing a portable Slip Simulator at the LANL training center. Two permanent Slip Simulators were installed in April of 2011. Two classes are offered and tracked for the Slip Simulator Experience; one is for workers who just observe the class and another for workers who choose to get on the simulator as a participant. The results of the two different populations are compared for the subsequent effectiveness of each.

As of 3/1/12, 2,562 total workers at LANL have completed either the observer or participant Slip Simulator training.

- 913 Participants received verbal instruction and practical instruction on the Slip Simulator (Participant training).
- 1,649 Observers received verbal instruction and watched others on the Slip Simulator (Observer training).

Injury cases involving slipping on slick surfaces at LANL were reviewed for a 12 month period from March 2011 to February 2012. The cases include all visits to Occupational Medicine as a result a “Contributing Factor” or Slip/Trip/Fall, regardless of the accident severity. 147 total Slip/Trip/Fall cases were identified, of which 62 involved falls on ice, snow or other slick surfaces, such as a wet floor.

Of the 913 employees with Participant training on the Slip Simulator:

- 3 fell on a slick surface BEFORE receiving the Participant training
- 0 fell on a slick surface AFTER receiving the Participant training

Of the 1,649 employees with Observer training on the Slip Simulator:

- 1 fell on a slick surface BEFORE receiving the Observer training
- 8 fell on a slick surface AFTER receiving the Observer training

Of the roughly 11,000 workers at LANL, during the 12 month period starting in March of 2011, 62 (0.56%) workers fell on a slick surface. By comparison, 8 (0.48%) of the employees who received Observer training fell on a slick surface after receiving the training. The data seems to indicate that there is minimal benefit from the Observer training, but significant benefit from the Participant training.

SCIENTIFIC REPORT

1. Background

The causes of occupational slip and fall accidents, both in terms of extrinsic and intrinsic factors and their associations are not yet fully understood. Successful intervention solutions for reducing slip and fall accidents require a more complete understanding of the mechanisms involved. Before effective fall prevention strategies can be put into practice, it is important to observe the chain of events in an accident, comprising the exposure to hazards, initiation of events and the final outcome leading to injury and disability. These events can be effectively identified and analyzed by applying epidemiological, biomechanical and tribological research principles and methodologies.

I. Epidemiology of Occupational Slips and Falls

Reducing slip and fall accidents has been the goal of many researchers since the 1920s. Although much has been learned over the last few decades about tribometric techniques to assess shoe/floor interactions, biomechanical responses to walking on slippery floor surfaces, and postural control, fall accidents continue to represent a significant burden to industries, both in terms of human suffering and economic losses. Older workers are particularly at risk. Many studies have shown that with advancing age, there is an increasing incidence of slip and fall injuries, and same level falls were the most common type of incident among older workers (Layne and Pollack, 2004). Falls are the second leading cause of work-related fatalities next to motor vehicle accidents, and the number of fatal falls exceeds the combined number of workplace deaths associated with poison, electric current, fire, burns and drowning (Leamon and Murphy, 1995). Foot slippage is the most frequent unforeseen trigger event for falls on the same level and falls to a lower level (Andersson & Lagerlof, 1983, Courtney et al., 2001). Gronqvist and Roine (1989), in a survey of occupational accidents, indicated that the majority of falls to a lower level were caused by slipping. In 1998, the U.S. Bureau of Labor Statistics reported that falls accounted for 16.8% of all non-fatal injuries involving days away from work and 11.9% of job-related deaths (BLS, 1998). According to the Bureau of Labor Statistics (2004), floors, walkways or ground surfaces were the major sources of fall accidents, causing over 86% of all fall-related injuries. Additionally, intrinsic factors, such as occupationally induced localized muscle fatigue, are considered as major factors contributing to slip and fall accidents (Davis 1983, Bentley and Haslam, 2001, Cohen and Lin, 1991, Maiti et al., 2001, Gauchard et al., 2001, Hsiao and Simeonov, 2001).

A review of literature on occupational falls revealed that slipping and tripping are often the “trigger events” for the fall accidents (Andersson and Lagerlof, 1983, Manning et al., 1988, Courtney et al., 2003). An epidemiological study (Cohen and Compton, 1982) characterizing the most frequent and serious work related injury in various industries, tasks, and work surface conditions suggested that the construction trades (e.g., painters, roofers, sheet metal workers) have a higher percentage of injuries associated with falls to a different level, while, office and service workers, restaurants, retail stores, hospitals, clothing manufacturers, and meat processing plants have a higher percentage of injuries associated with falls on the same level. Similar findings were reported by Manning et al., (1988), in a study of underfoot accidents in a work population of 10,000 in 1985. They reported that the majority of events prior to injury (62%) was slipping. Similar findings were also reported by Davis (1983), Tisserand (1985), and more recently by Courtney et al., (2003).

In terms of occupational injury and illness due to slips and falls, the BLS (1997) estimated that there were 333,913 non-fatal, fall-related injures in 1996 involving one or more days-away- from-work. A majority of these injuries were attributed to falls on the same level (66%) and falls to a lower level (30%) (Courtney et al., 2001). However, the majority of fall fatalities occurred as a result of falls to a lower level. Overall, there were approximately 390,241 disabling occupational injuries due to slips and falls (including the bodily reaction event category associated with slips and trips without falls). Furthermore, data from US National Electronic Injury Surveillance System indicated that the number of occupational fall-related injuries treated in U.S. emergency departments in 1998 was 550,592, with an additional 49,661 injuries resulting from slips and trips and losses of balance without a fall. These injuries accounted for 48% of disabling sprains and strains and 46% of disabling fractures (Courtney et al., 2001). The distribution of fall-related days-away-from-work occupational injures were similar for men and women, and falls on the same level had median days-away-from-work of 7 days (higher than median days-away-from-work of 5 days for all occupational injury cases).

In summary, a majority of occupational falls leading to injuries and deaths are a result of foot slippage and are typically experienced by unskilled and skilled laborers. Although government, labor, and industry organizations have been working to reduce the risks of fall-related injuries through stepped-up inspections, monitoring of worksite conditions, and through comprehensive safety training for many years, workers are continuously faced with the potential harms associated with fall accidents. These findings warrant the need for additional studies to provide more effective prevention strategies and design criteria for jobs and working environments to reduce occupational slip and fall accidents.

II. Mechanisms of Slips and Falls

A review of literature on occupational fall accidents indicated that multiple mechanisms are involved in slips and falls. Numerous studies have identified work related risk factors for falling. Factors intrinsic and extrinsic to the worker, and the hazards and demands of the environment, contribute to most falls in varying degrees. In general, the ability to walk safely and preserve balance (keeping the body's center-of-mass [COM] over its base of support) in the event of a slip and fall is dependent upon intact sensory and musculoskeletal systems. However, with advancing age and muscular fatigue, a variety of physiologic changes affecting these systems may interfere with gait and balance, placing these individuals at a higher risk for slip and fall accidents.

Effects of Localized Muscle Fatigue on Slip-Induced Falls

Modern technology has contributed greatly to the reduction of heavy work in industrialized countries, however, labor demanding intense physical work is still necessary in some occupations such as construction, forestry and many service occupations (Astrand and Rodahl, 1986). It has been estimated that a third of the U.S. workforce must exert significant strength as part of their jobs, and experience fatigue at work places (Swaen et al., 2003). In a healthy person, fatigue is a normal phenomenon, experienced by everyone and usually relieved by rest. Localized muscle fatigue has been defined as the inability of the muscles to maintain expected force output or a reduction in the force generating capacity of the total neuromuscular system (Vollestad, 1997). The concept of localized muscle fatigue as a precursor to accidents is not new and has been successfully applied to avoid musculoskeletal disorders (i.e., recommended lifting capacity). In the context of fall accidents, the literature (Lipscomb, 2006, Hsiao and Siemonov, 2001) provides convincing arguments that localized muscle fatigue can disrupt the quality of the signal from the periphery for effective balance control during slip perturbation and increase the risk of slips and falls.

Fatigue is a complex phenomenon which is widely known but difficult to quantify. It can be brought about by a person's motivation level, a build-up of metabolites (e.g., lactic acid) in the muscle, a loss of energy supply, or a combination of the above. Fitts (1996) defined muscle fatigue at a cellular level as a "decrease in the peak tension and power output resulting in a reduced work capacity, depending on a person's state of fitness, muscle fiber type and composition, and type of exercises being performed." During heavy dynamic work or static exercise, the blood circulation cannot keep up with muscular demands for oxygen supply and lactate removal which results in reduced endurance and decrements in contractile capacity and motor precision (Kilbom 1990). There is a considerable controversy about how fatigue is defined and measured. It is argued that before the failure point is reached, the muscle is already fatigued (De Luca, 1984). Fatigue, from this perspective, is an ongoing and gradual process that begins at the start of a muscle contraction, rather than an abrupt event. For the purpose of this study, fatigue is defined as a decrease in the maximum force-producing ability.

Often, the fatigability of a muscle is characterized by either the time that the required force or power output can be sustained (endurance time), or the extent that force or power are reduced in a given time period (Allman and Rice, 2002). Many researchers, however, use operational definitions dependent on measurement method to describe fatigue, such as increased electromyography (EMG) activity, shift of EMG power spectrum towards low frequencies, and impaired force generation (Oberg et al. 1994). The shift in the EMG spectral density towards lower frequencies is correlated with a reduction in propagation velocity of the action potential along the muscle fibers (Lindsrom et al., 1970). During repetitive, isometric contractions, the recruitment pattern of the motorneurons can occur according to the size principle (i.e., small units are activated at low forces). As a result, mechanical strain is placed on a few motor units because the same units can be recruited continuously during a given work task (Sejersted et al.,1993). Fatigue and incomplete recovery of these motor units could trigger biomechanical damage. Numerous variables, including postural stability, maximum voluntary contraction force, and reaction time, have been studied in conjunction with fatigue protocols to help understand how fatigue affects the body and the ability of the body to perform work tasks.

The onset of localized muscular fatigue is detrimental to productivity and performance of workers due to alterations in various physiological and biomechanical characteristics. Accumulation of metabolic by-products reduces the intramuscular conduction velocity until a point where the muscles are unable to produce the desired force output (Gefen et al., 2002). Existing literature provides support that localized muscle fatigue adversely affects proprioception (Skinner 1986), movement co-ordinations (Sparto, 1997), and muscle reaction times (Hakkinen, 1986). Armstrong (1993) described further mechanical and physiological changes that occur in muscles as they fatigue, such as deformation and yielding of connective tissue within the muscle and increases in intramuscular pressure. Other processes such as ion shifts, electrical excitation, and shifting concentrations of substrates and metabolites also occur (Cady et al., 1989). These changes are conveyed to the central nervous system, causing sensations of effort and discomfort (perceived fatigue).

The time to develop an injury and the extent of injury depend on the amount of recovery time between periods of fatigue. If the activities are such that the muscle is unable to adapt to the exertion and are repeated over extended periods of time, joint and tissue degeneration and inflammation and chronic pain can develop (Chaffin et al., 1991). A muscle's rate of recovery from fatigue depends on several factors, including duration of a work task, the intensity of a work task, and the physical fitness of an individual (Armstrong, 1993). Therefore these factors can be utilized while designing efficient work-rest schedules in various work settings.

Fatigue can be measured subjectively through the use of scales or questionnaires or objectively through physiological methods. Some of the objective measures include changes in posture, altered muscle coordination, and changes in performance accuracy. There has been ongoing debate about which of the measures are the most valid indicators of fatigue. Although there has been no definitive solution to this problem, most of the valid measurements of fatigue depend on how fatigue is defined and on the reliability of measures. As fatigue in this study is defined as a decrease in maximal ability, it can be measured directly by taking maximum voluntary contractions (MVC) measurements over the course of dynamic activity. The value of the participant's MVC is considered a "gold standard" for identification of fatigue occurrence (Vollestad, 1997). MVC is defined as the maximum force generated by a participant when the participant is encouraged to perform a contraction to their highest ability.

III. Risks of Falls Among Older Workers

The age distribution and the mean age of the labor force are undergoing rapid and significant changes worldwide. In the United States, conservative projections of population demographics indicate that the number of older workers (labor force comprised of those 45 years and older) will increase from 33% (1998) to 45% (2008) (Fullerton, 1999, Dohm, 2000), and by 2010, the median age of workers is projected to reach 40.6 years (Fullerton and Toossi, 2001). According to the US Census Bureau (2001), 59 million men and women in the civilian non-institutionalized population were aged 55 years and over and approximately 70% of 55 to 59 year olds, 50% of 60 to 64 year olds, and 15% of 65 and over were in the civilian labor force. These demographics clearly suggest that risks and prevention of work-related falls must be evaluated in high-risk groups such as the aging workforce in the U.S.

From a recent review of occupational fall accidents (NIOSH, 2000), older workers appeared to be at a higher risk of fall-related injury and death. Many studies have shown that with advancing age there is an increasing incidence of fatal slip and fall injuries (Campbell et al., 1981, Rubenstein et al., 1988, Rice et al., 1989). Falling occurs often among older individuals and ranks as the second highest cause of accidental death for 45-75 year olds and the highest accidental cause of death for those aged 75 years and over (National Safety Council, 2002). For older individuals, falls on the same level make up the largest percentage of accidents. While young people fall more frequently than older individuals, the injury rate, particularly for serious fall-related injuries, is higher among older adults (Layne & Pollack, 2004).

Major physiologic changes affecting the potential for slip and fall accidents begin to appear starting in the mid-twenties. In general, isometric muscle strength peaks in the mid-twenties and then decreases slowly until after 50 years of age when there is an accelerated decline (Rice & Cunningham 2001, Astrand & Rodahl, 1986, Larsson, 1982). These declines in strength development appear to stem from changes in muscle contraction mechanisms (Thelen et al., 1996), mitochondrial enzyme activity (Houmard et al., 1998), and in the proportion of fast-twitch to slow-twitch muscle fibers (Lexell, 1995). Studies suggest that age-related changes in muscle strength have an important effect on recovery of slip and fall accidents (Lockhart et al., 2005, Bonder and Wagner, 1994, Whipple et al., 1987, Larsson, 1979, Wolfson et al., 1985 and 1995). This effect can be further aggravated by fatigue and increase the risk of falls among older workers.

A number of studies have documented the decline of postural control (keeping the body's COM over its base of support during quiet stance and active movement) due to sensory degradation among older adults (Sheldon, 1963, Nashner, 1990). This decline of postural control is believed to be an integrative process associated with a greater risk of falling (Maki et al., 1996, Isaacs 1985). Sensory inputs important for balance are: vision, proprioception, and vestibular sensations (Lacour et al., 1983, Nashner, 1980). Vision plays a major role in maintaining stability, both in quiet stance and while undergoing movement such as walking (Tinetti & Speechley, 1989). Visual acuity, adaptation to the dark, peripheral vision, contrast sensitivity, and accommodation, all of which are related to stability, may be affected by age-related changes (Kornzweig, 1977, Goldman, 1986, Cohn & Lasley, 1985). For example, age-related decrements in peripheral vision may impair an older individual's use of visual reference information. Narrowing the visual field may deprive the older person of that part of the visual field most sensitive to movement (Stelmach & Worringham, 1985) and, as a result, postural control may be compromised. Studies indicate that proprioceptive deficits are also significantly higher in older individuals (Rabbitt & Rogers, 1965, et al., 1982 and Skinner et al., 1986). The proprioceptive system contributes to stability, particularly during changes of position. Woollacott et al., (1986) demonstrated the significance of ankle proprioception for balance retention in the elderly by utilizing a moveable platform that permitted either stabilization of the visual field or of the support surface. Comparing postural sway, they found that the elderly had large increases in postural sway when ankle proprioception was eliminated (via a moveable platform). Proprioceptive information also plays a vital role in modification of internal models using feedforward control mechanisms (Ghez & Sainburg, 1995, Bard et al., 1995). During a slip perturbation, motor programs have to be modified to maintain dynamic stability. Modification of the motor program is closely associated with visual input as well as proprioceptive input. However, in a situation where conflicting visual cues exist with the environment, proprioceptive input may be the quickest and most accurate modality associated with balance maintenance (Ghez & Sainburg, 1995). As a result, older adults' proprioceptive deficits may hinder optimum balance recovery during slip-induced perturbations and increase the likelihood of falls especially in a fatigued state.

Furthermore, increasing age can also have an effect on gait due to postural and balance changes. Older adults tend to have a shorter step length and a broader walking base, which results in an increase in stance time and double support time (Murray et al., 1969, Gillis et al., 1986, Imms & Edholm 1979, Winter et al., 1990). Many researchers have observed that on slippery floor surfaces, subjects tend to shorten their step length in order to reduce foot velocities, foot shear forces, and reduce the likelihood of slipping (Cooper & Glassow, 1963, Ekkubus & Killy, 1973). As a result, the shorter step length and the broader walking base of older adults are thought to result in a more stable or safer gait pattern. However, these gait adaptations may have some important implications for the initiation of slip-induced falls. As indicated earlier, Winter et al., (1990) and Lockhart (1997) reported that the horizontal heel velocity during the heel contact phase of the gait cycle was significantly higher for older subjects than for younger subjects, even though the walking velocity of older subjects was slower. This increase in horizontal heel velocity during a critical time of weight transfer might increase the potential for slip-induced falls as discussed earlier. Thus, general gait instability among older individuals, and specifically higher horizontal heel velocity during the critical phase of the gait cycle, may increase the friction demand thereby increasing the likelihood of slip-induced fall accidents. This effect can be further aggravated by fatigue (Saggini et al., 1998) and increase the risk of falls among older workers.

In summary, gait instability, sensory degradation, and diminished rapid torque development capacities of the older workers imply that age must be considered as a factor in the identification of risk of occupational falls.

IV. Other Intrinsic and Extrinsic Factors Related to Occupational Slips and Falls

Inclined Support Surface

Walking down a ramp/roof poses a significant hazard to workers due to the generation of higher shear forces when ambulating over an inclined surface than a level surface. Bentley & Haslam (2001) indicated that not only roofers but also postal workers walked on downward slopes leading to 30% of slip and fall accidents. Ground reaction forces have been investigated during descent of a ramp (McVay & Redfern, 1994, and Harper et al., 1967). These investigations suggest that shear force increases as ramp angle is increased. Thus, the RCOF (i.e., friction demand) at the shoe/floor interface can increase with increases in ramp angle. For example, McVay & Redfern (1994) found that the mean of the peak RCOF across subjects increased from about 0.25 to 0.5 as ramp angle is increased from 0 to 20 degrees at heel contact and throughout the gait cycle. In order to prevent slips and falls, working on steep roof surfaces may require high slip-resistance characteristics for the shoe/floor interface, and above certain slope, slip-resistance required for walking will not be possible to achieve. Although theoretical and experimental evidence provides support for an increased risk of slipping while ambulating on inclined support surfaces, the threshold level of the angle of roof inclination and effective shoe/floor interfaces for safe walking has not been scientifically determined.

Work Pace (Walking Speed)

Natural or free cadence is defined as the steps/min when a subject walks at self-selected speed and ranges from 101 – 122 steps/min for adults less than 65 years old (Winter, 1991). When rushed, typical industrial tasks require workers to perform at a greater work pace than normal walking pace. When an individual walks with a faster cadence (e.g., rushed), one must override their natural frequency and consciously force cadence to a faster rate to increase the walking speed. Walking speed directly affects the magnitude of shear force (F_h), and therefore also has a direct effect on the friction demand (RCOF) (Kim, Lockhart, and Yoon, 2005). An increase in walking velocity usually increases the friction demand and risk of slip initiation (Soames and Richardson, 1985). Dynamic stability of walking is also influenced by walking velocity. Dingwell et al., (2001) observed that increased walking speed reduces dynamic walking stability. Reduced dynamic stability of walking may be related to two factors including dynamics of energy balance and speed of neuromuscular control. Neuromuscular response must be faster at greater walking velocities to accommodate the quicker time sequences of fast walking. This may be related to momentum, i.e., a perturbation or error at high velocity has greater momentum than at low velocity, and requires a larger neuromuscular response to correct and stabilize the system. As such, faster work pace or walking speed during rushed industrial activities may adversely affect slip initiation and fall recovery processes. Although theoretical and experimental evidence provides support for an increased risk of slipping and falling while walking at faster pace, the relationship between muscle fatigue, walking speed, and risk of falls has not been scientifically determined.

Load Carrying

In normal walking, corrective postural movements are made by the upper body, arms, and shoulders. Arm swing is used to offset the rhythmical acceleration and deceleration of the trunk by the leg movements, and also to damp-out the rotational forces of the trunk arising from the same causes (Haywood, 1986). However, these dampening effects are modified during the laden state (Davis, 1983) and alter gait and posture and may influence risk of slip initiation (Liu and Lockhart, 2006). Load- carrying (in front) also displaces the whole-body COM anteriorly, placing it closer to the forward edge of the supporting base requiring additional rotational torque at the foot-ground contact. Hsiang and Chang (2002), utilizing an inverted pendulum model demonstrated that when load was carried in front of the whole-body COM, the added load increased momentum pulling the COM forward into the next step. Subsequently, less active force was needed during the push-off phase of the gait cycle. Reduced push-off force of the stance leg will reduce the transitional acceleration of the whole-body COM and increase friction demand (Lockhart et al., 2003) indicating a potential increase in slip initiation risk. However, gait accommodations are necessary to walk while keeping the whole-body COM within its stability limits during laden walking (Pai and Patton, 1997). This compensatory behavior may be related to efforts to optimize the gait pattern for energy efficiency, maximum speed, or other criteria such as stability while carrying a load (Collins, 1995; Liu and Lockhart, 2007). This compensatory walking behavior has been documented and is influenced by weight of the load and speed of walking. For example, Cham and Redfern (2004) investigated the effects of load-carrying on gait characteristics and friction demand. These investigators used light loads (i.e., no-load, 2.3kg and 6.8kg) and suggested that load-carrying was associated with small but significant **decreases** in the required friction properties for safe walking (i.e., less slip propensity attributed to **increased knee and hip flexion**, shorter step length and slower walking velocity). Similar results are reported by Kim and Lockhart (in press) carrying a load equivalent to 10% of body weight. **Conversely**, when carrying a heavier load, Myung and Smith (1997) observed higher heel-contact velocity, friction demand, and consequently, an increased risk of slip initiation. However, protocol differences limit the comparison between these studies. For example, participants in Cham and Redfern's study walked with natural cadence, and participants in the Myung and Smith study walked at a fixed pace of 1.33 m/s. This discrepancy must be investigated in controlled experimental trials (e.g., using a metronome to set a walking speed with different load levels). Furthermore, although theoretical and empirical evidence provides support for the alteration of slip propensity and risk of falling while carrying a load, the relationship between load-carrying, muscle fatigue (especially the joint effects), and risk of slip-induced falls has not been scientifically determined.

The Need for Research on Fall Prevention

As noted by NIOSH (2000), “exposure to fall hazards is a nearly constant aspect of employment.” As such, the accident statistics presented earlier may remain unchanged unless substantial progress is made towards new means of reducing occupational fall occurrences. Such reductions are typically achieved through either fall protection (for fall to a lower level) or fall prevention (Hsiao and Simeonov, 2001). Fall protection refers to interventions aimed at minimizing injury severity after a fall event is initiated. The current OSHA (1995) standard for fall protection and other existing standards address this issue, covering fall protection systems, guardrails, and other related approaches. Despite the existence of such standards, such protection systems are not currently used with sufficient frequency or correctness (NIOSH, 2000). Furthermore, fall protection can not prevent an incident from occurring (Smith and Veazie, 1998). Given this evidence, reliance on fall protection may not be an effective approach towards minimizing occupational fall accidents. Furthermore, although a static coefficient of friction of 0.5 is recommended to prevent a slip on a level surface (Redfern and Bidanda, 1994, Lin et al., 1995), **currently no standard or law requires that a floor must have a certain level of slip resistance.** Additionally, the frictional properties of support/walking surfaces (e.g., roofing materials - asphalt shingles and roofing rolls, etc.) are not well defined and may change significantly with changes in environmental conditions (e.g., heat and moisture) and hamper balance control during dynamic tasks (Chang et al., 2001, and Hsiao & Simeonov, 2001). Given this evidence, reliance only on floor COF recommendation may not be an effective approach towards minimizing occupational fall accidents. Rather, fall prevention remains an important and perhaps even a critical task. As such, there is an important need for methods to evaluate existing and new prevention strategies, and more generally to determine factors that lead to occupational fall accidents. Since most occupational falls appear to be initiated by foot slippage, fall prevention can be facilitated by a better understanding of friction demand characteristics (i.e., RCOF) while ambulating on level and inclined surfaces at normal and faster walking speeds (with and without a load), factors that adversely affect slip initiation and recovery (fatigue and aging), and interventions that promote safe walking.

In summary, occupational fall accidents remain a significant cause of injuries and fatalities among the American workforce. A majority of these incidents are a result of foot slippage experienced mostly by laborers, and fatigue and aging appear to introduce additional risk of falls. Although mechanisms and national standards for fall protection exist, personal protective equipment is most often not used or is used incorrectly, and fall protection only minimizes the severity of fall outcomes. Additionally, human interactions to frictional properties of support/walking surfaces are not well understood. Improved fall prevention is therefore necessary to move towards a reduction of fatalities and injuries related to occupational slips and falls.

This proposal addresses this need through a combination of experimental studies and biomechanical modeling. The proposed work will provide a better understanding of fatigue and aging as factors that contribute to slip-induced falls, will generate a quantitative mechanism to evaluate existing and new intervention strategies for improving balance during fatiguing work. The proposed work addresses several NORA Priority Areas in the context of work-related traumatic falls. The main Priority Areas are: 1) Risk Assessment Methods; 2) Control Technology; and 3) Intervention Effectiveness Research.

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2. Specific Aims

In summary, localized muscular fatigue can influence the initiation, detection, and recovery phases of slips and falls and may increase the likelihood of work-related slip-induced fall accidents. If localized muscular fatigue affects the initial gait characteristics, the potential for slip-induced falls may increase. Furthermore, there are certain processing stages that must be undertaken during the detection phase if a fall is to be avoided or compensated for (recovery phase). During the detection phase, if a potential fall is imminent, sensory input must trigger or alert those centers responsible for response selection. As the literature provides convincing arguments that localized muscular fatigue can disrupt the quality of the signal from the periphery for effective balance control and may delay the response selection-execution increasing the risk of occupational falls. Additionally, inability to generate the necessary counterbalancing joint moments due to fatigue during recovery either in magnitude or in rate of development to control the body's horizontal and vertical momentum can increase the risk of falls.

Specific Aim: To quantify the effects of localized muscle fatigue on slip propensity and fall recovery and balance control strategies during slip perturbations. Fatigue (as measured by a reduction in force output) was induced in lower extremity joints (ankles, knees, and combination of ankles, knees and hips) and low back to assess the effects of fatigue location on slip propensity and fall recovery control strategies. Additionally, the differential effects of work pace, age, load carriage, and level of floor inclination was determined. The sensitivity of several common measures of slip propensity/slip severity will also be assessed. Two separate experiments was conducted to evaluate slip-initiation and balance-recovery characteristics while walking over non-slippery and unexpected slippery floor surfaces.

Primary Hypotheses

1. Localized muscle fatigue will adversely affect slip initiation and balance recovery.
2. Fatigue-related decrements (slip and fall propensity) will differ in severity among fatigue locations, age groups, floor inclined levels, load levels, and work pace conditions.
3. There will be fatigue- and age-related differences in joint torque and power, estimated using an inverse dynamic model during slip perturbations.

Specific Aims concerning the "slip-initiation" characteristics are presented below discussing the gait changes associated with lower leg localized muscle fatigue and, slip propensity changes associated with two different types of lower extremity muscle fatigue conditions (i.e., squat and leg extensions on the Biodex). Additionally, most recent results regarding the ankle fatigue condition are introduced here. Future researches concerning the slip initiation characteristics are further elaborated. Afterwards, the effects of fatigue on slip recovery characteristics are further explored. Here, two studies are discussed in lieu of fall recovery responses to a slip perturbation associated with quadriceps fatigue and back fatigue conditions.

3. Background: Epidemiology of Occupational Slips and Falls

Yeoh, H., Lockhart, T.E., and Wu, X., (2013), Non-Fatal Occupational Falls on the Same Level. *Ergonomics*, 56,2: 153-165.

Abstract

The purpose of this study was to describe antecedents and characteristics of same level fall injuries. Fall incidents and costs were compiled from the Bureau of Labor Statistics and other sources from 2006–2010. This study indicated that over 29% of 'fall on same level' injuries resulted in 31 or more workdays lost. The major source of injury was 'floors, walkways or ground surfaces', and the most affected body parts were the lower extremities and the trunk. With regard to gender and age, female workers had the highest risk of falls, while advancing age coincided with an increase in incidence rates. Overall, workers in the healthcare and social assistance industry, the transportation and warehousing industry, and the accommodation and food services industry had the highest risk for 'fall on same level' injuries. Furthermore, the overall compensation cost increased by 25% from 2006–2009. Along with existing evidence, these results may facilitate the design and implementation of preventative measures in the workplace and potentially reduce fall-related compensation costs.

Practitioner Summary: This research presents a unique and detailed analysis of non-fatal 'fall on same level' injuries in a large population of workers from various private industries in the USA. This information can be used to prioritise designing and implementing preventive measures and to provide workers with the understanding of risk factors associated with falls in the workplace.

Keywords: falls; fall on same level; occupational injuries; characteristic of injured workers; consequences of occupational fall

Introduction

Injuries associated with fall accidents pose a considerable threat to the USA in terms of both human suffering and economic losses. According to the Bureau of Labor Statistics (BLS), the three most frequent fatal injuries are highway incidents, falls and homicide (BLS 2010a). Approximately, 9 out of 10 non-fatal occupational injuries result from three events: bodily reaction and exertion, contact with objects and equipment, and 'fall on same level' (BLS 2008). For the first time in the history of the industrialised world, the combined cost of all fall-related occupational injuries surpassed overexertion injuries caused by excessive lifting, pushing, pulling, holding or throwing an object with a significant burden on our economic system (National Safety Council 2006) and are responsible for a significant proportion of worker absenteeism (Courtney et al. 2001; Courtney and Webster 2001). The annual direct cost of occupational injuries due to falls in the USA is expected to exceed $43.8 billion by the year 2020 (Englander, Hodson, and Terregrossa,1996).

Although the elements behind occupational fall accidents are not yet fully understood, a number of demographic, lifestyle and workplace factors have been linked with risk of injury in an occupational accident (Swaen et al. 2004; Laflamme, Menckel, and Lundholm 1996; Chipman 1995; Frone 1998; Wells and Macdonald 1999; Leistikow et al. 2000; Frank 2000; Nag and Patel 1998). Successful interventions for reducing falls require a comprehensive knowledge of the mechanisms involved. This study describes the characteristics of occupational fall accidents, the initiating events and the final outcome leading to injury and disability. Moreover, because of the growth trends in occupational 'fall on same level' incidents, the non-fatal occupational injury category related to this event is of particular interest. Hence, the objective is todescribe occupational 'fall on same level' injuries in US private industries, categorised by the following factors: major US industry, nature of injury, source of injury, types of exposures or events, part of body injured, occupation, age of injured, gender of injured, the number of days away from work (DAFW) due to fall related injuries, time of fall occurred and, finally,the cost of these injuries.

Methods

Information and costs accompanying non-fatal incidents were compiled from two sources: the US BLS and the Liberty Mutual Research Institute, Workplace Safety Index (WSI). The data we accessed are available at the BLS (2012) and Liberty Mutual Research Institute website (Liberty Mutual 2010).

The US BLS national data

The BLS compiles national data on non-fatal occupational injuries and illnesses in the private industry from the Survey of Occupational Injuries and Illness (SOII) and estimates the overall occupational injury and illness experience (BLS 2011). The BLS categorises four main events or exposures in fall-related injuries: 'fall to lower level', 'fall on same level', 'jump to lower level' and 'unspecified' events. 'Fall on same level' events occur when contact with the source of injury is made on the same level or above the surface supporting the injured person. 'Fall to lower level' events transpire when the source of injury makes contact below the surface level supporting the individual. Conversely, 'jump to lower level' events arise when the injured person voluntarily leaps from an elevation, albeit to avoid an uncontrolled fall or other injury. Events peripheral to these categories are labelled 'Unspecified'. To estimate the impact of 'fall on same level', and consequently overall occupational fall injuries in the US, 'slip, trip, loss of balance without fall' injuries were also included. 'Slip, trip, loss of balance without fall' injuries were classified under the bodily reaction event category, where a worker slipped, stumbled, mis-stepped but did not fall. (The title was switched to OIICS 2.0 after the coding structure underwent a comprehensive revision in 2010. Under this new structure, the BLS moved 'slip, trip, loss of balance without fall' events out of the overexertion category and into a new category titled, 'slips, trips and falls'.) 'Slip, trip, loss of balance without fall' was estimated by using BLS' yearly tables entitled 'Table R4'. Non-fatal injuries and illness-related work absences are classified by nature, source, injured body part, age, gender, occupation, race, time when injury occurred and length of service involving one or more DAFW. Agricultural establishments with fewer than 11 employees, self-employed individuals and federal government employees are excludedfrom the survey. This study is a compendium of fall-related injury data based on the SOII, between the years 2006 and

2010. The BLS tables were examined to extract physiologically meaningful injury statistics relevant to occupational falls and toprovide the number of incidence, percentage (%) and incidence rate, where incidence rate embodies the number of injuriesper 10,000 full-time workers. Specifically, the number of cases and per cent distribution of fall injuries involving DAFW were extracted from the BLS yearly supplemental table entitled 'TABLE 3 Number, percent distribution, and median days away from work for nonfatal occupational injuries and illnesses involving days away from work by selected worker and case characteristics and falls.' Incidence rates for 'fall on same level' injury, part of body, source, occupation, age group and gender were estimated from tables R8, R24, R30, R100, R110 and R111, respectively, for each studied year. Furthermore, per cent distribution of number of DAFW, time of day injury occurred and affected industry were mined from tables R70, R90 and R113, respectively.

The Liberty Mutual Research Institute worker compensation data

The worker compensation costs of fall-related injuries were derived using the Liberty Mutual Research Institute, WSI. The WSI consolidates serious non-fatal workplace injuries and identifies the explicit causes behind the events. WSI was developed by applying annual Liberty Mutual workers' compensation claims cost data to the workplace accident frequency information provided by the US Department of Labor's Bureau of Labor Statistics. The data were then applied to national estimates of the cost of workers' compensation benefits from The National Academy of Social Insurance, which includes information from a broad range of workers' compensation insurance companies. Using injury event definitions developed by the BLS, the Liberty Mutual Research Institute amasses injury data associated with employees missing 6 or more days from work, and ranks them by total worker compensation costs (Liberty Mutual 2010). The information obtained from BLS tables were used in conjunction with the workers' compensation cost to identify as many specific details as possible about the circumstance of 'fall on same level' events to further delineate conditions that might have contributed to these injuries.

Results

Table 1 specifies the 5-year period from 2006 to 2010, where approximately 5.32 million work-related non-fatal injuriesinvolving DAFW occurred in the USA; 21.6% (1.14 million) of these accompanied falls. Subsequently, 'slip, trip, lossof balance without fall' induced 170,270 injuries, contributing to around 3.2% of overall annual private industry injuries.The percentage of fall-related injuries classified by four main events involving DAFW from 2006 to 2010 is illustrated in Figure 1. Table 2 encompasses occupational injuries, overall fall injuries and fall-related injuries involving DAFW in the private industry. Among the fall events, 'fall on same level' cases correlated with the most fall-related injuries, including an average of 66.2% of total occupational falls. 'Fall to lower level' cases were a distant second, with an average of 29.8%. Workers during this period were two times more at risk of 'fall on same level' injuries than 'fall to lower level' injuries, with an average incident rate of 16.5-7.4 over the 5-year period.

Table 3(A) shows that female workers experienced a greater distribution of 'fall on same level' injuries involving DAFW compared to their male counterparts (55.7% female; 44.3% male). In particular, female employees were two timesmore at risk than male employees over the 5-year period, with incidence rates of 22.0 and 12.5, respectively.

Table 1. Number, percentage and incidence rate of occupational injuries involving DAFW by selected events in private industry for all United States, 2006–2010.

All events		All fall[a]			Fall on same level			Slip or trip without fall		
Year	Number	Number	Percentage[b]	Incidence rate[c]	Number	Percentage[b]	Incidence rate[c]	Number	Percentage[b]	Incidence rate[c]
2006	1,183,50	234,450	19.8	25.3	151,750	12.8	16.4	35,440	3.0	3.8
2007	1,158,87	253,440	21.9	26.7	166,560	14.4	17.6	37,780	3.3	4.0
2008	1,078,14	234,840	21.8	24.7	157,680	14.6	16.6	35,420	3.3	3.7
2009	964,990	212,760	22.0	23.4	141,120	14.6	15.6	32,490	3.4	3.6
20010	933,200	208,470	22.3	24.1	139,660	15.0	16.1	29,140	3.1	3.4

Source: Bureau of Labor Statistics (2012).
[a] Includes fall to lower level, fall on same level and jump to lower level. [b] Percentage of overall non-fatal occupational injuries in private sectors. [c] The number of injuries per 10,000 workers.

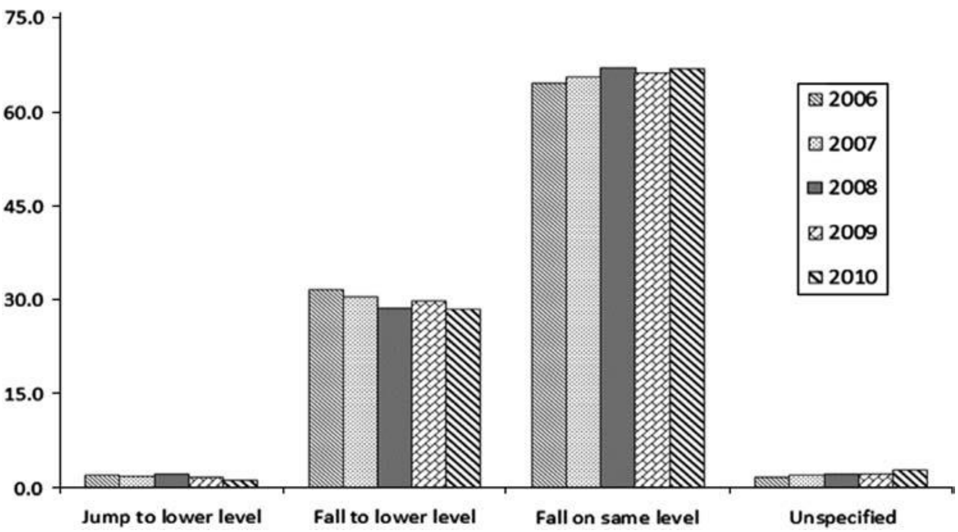

Figure 1. Percentage of occupational injuries by fall-related events in private industry. *Source*: Bureau of Labor Statistics (2012).

Table 2. Number, percentage and incidence rate of occupational injuries involving DAFW by falls in private industry, 2006–2010.

	All fall[a]	Fall on same level			Fall to lower level			Jump to lower level		
Year	Numbe r	Number	Percentage[b]	Incidence rate[c]	Number	Percentage[b]	Incidence rate[c]	Number	Percentage[b]	Incidence rate[c]
2006	234,45	151,750	64.7	16.4	74,280	31.7	8.0	4590	2.0	0.5
2007	253,44	166,560	65.7	17.6	77,300	30.5	8.1	4560	1.8	0.5
2008	234,84	157,680	67.1	16.6	67,510	28.7	7.1	4830	2.1	0.5
2009	212,76	141,120	66.3	15.6	63,320	29.8	7.0	3670	1.7	0.4
2010	208,47 0	139,660	67.0	16.1	59,440	28.5	6.9	2580	1.2	0.3

Note: Slip or trip without fall data are not included, incidence rate embodies the number of injuries per 10,000 full-time workers.
Source: Bureau of Labor Statistics (2012).
[a] Includes fall to lower level, fall on same level and jump to lower level.
[b] Percentage of non-fatal occupational fall injuries in private sectors.
[c] The number of injuries per 10,000 workers.

In Table 3(B) workers between 45 and 54 years of age had the most fall-related injuries involving DAFW, with an average of 27.2%. The 55 – 64 and 35 – 44 age groups were the second and third largest fall populations with an average of 21% and 20.3%, respectively. The combination of these groups, ages 45 – 64 years, gave a total of 48.2% injuries associated with 'fall on same level' accidents and indicates that the incidence rate of falls increases with advancing age. The incidence rate for falls was highest among persons of 65 years and older with 34 falls per 10,000 workers, compared to the younger age group of 25-34 years consisting of 11 falls per 10,000 full-time workers.

Table 3. Characteristics of injured workers – same level falls involving DAFW in private industry, 2006 – 2010.

	2006		2007		2008		2009		2010	
Characteristics of injured workers	Percentage	Rate	Percentage	Rate	Percentage	Rate	Percentage	Rate	Percentage	Rate
(A) Gender										
Women	54.1	21.6	53.9	23.1	56.8	22.6	56.7	20.8	57.1	21.7
Men	45.9	12.8	46.1	13.8	43.2	12.3	43.3	11.7	42.8	12.1
(B) Age group (years)										
16–19	2.8	14.3	2.5	14.4	2.5	14.3	1.9	12.1	2.2	16.2
20–24	8.0	12.9	7.0	12.4	6.5	11.1	6.7	11.3	6.0	10.7
25–34	16.8	11.7	16.5	12.4	16.1	11.5	15.4	10.3	15.1	10.4
35–44	22.2	14.5	20.9	15.1	20.5	14.1	19.9	13.1	17.9	12.6
45–54	25.9	18.0	26.9	20.1	27.0	18.8	27.2	17.5	29.0	19.6
55–64	19.2	26.3	20.8	29.6	21.5	27.3	21.6	24.4	21.7	24.9
65 and over	5.2	33.1	5.5	34.6	6.0	33.7	7.2	36.1	7.0	34.6
(C) Occupation[a]										
Building and grounds cleaning and maintenance	6.6	36.4	7.1	42.2	7.0	39.4	7.9	42.4	7.2	39.7
Construction and extraction	7.8	21.5	7.2	21.5	6.1	16.9	5.5	16.8	4.8	16.8
Farming, fishing and forestry	1.0	16.6	0.8	15.1	0.7	12.4	0.9	14.0	0.9	14.1
Food preparation and serving	11.5	24.8	10.2	23.8	9.8	21.4	9.9	19.8	10.9	22.0
Office and administrative	10.1	10	10.5	11.1	11.5	11.5	9.8	9.2	10.4	10.2
Healthcare support	6.3	39.1	6.7	43.6	7.3	42.4	8.4	42.8	8.0	40.6
Installation, maintenance and repair	4.4	15.0	4.8	17.6	4.8	16.9	5.1	16.9	5.0	17.4
Production	9.6	16.1	9.4	17.2	9.0	16.1	7.8	14.1	8.4	16.3
Protective service	1.3	22.7	1.5	27.6	1.6	28.0	1.6	24.0	1.3	21.2
Sales	9.0	12.6	9.5	14.3	9.5	13.4	9.4	12.4	7.9	10.8
Transportation and material moving	15.6	29.8	15.4	32.4	14.9	30.2	14.4	28.6	15.2	31.6
(D) Industry[a]										
Accommodation and food services	-	25.5	21.8	22.1	23.4	21.2	20.3	19.5	20.7	21.3
Agriculture, forestry, fishing and hunting	-	22.1	10.1	18.2	9.3	16.9	10.5	17.1	12.4	20.4
Construction	-	17.5	9.3	17.7	7.6	13.3	8.8	13.8	9.6	14.3
Healthcare and social assistance	-	27.0	20.2	29	20.4	28.5	20.7	28.6	18.9	26.5
Manufacturing	-	12.5	9.3	13.3	7.6	12.7	8.8	10.6	9.6	11.5
Mining	-	16.9	10	13.9	8.5	10.9	9.8	10.6	11.3	11.5
Retail trade	-	19.1	16.2	22.2	16.1	19.4	15.2	18	14.2	16.8
Transportation and warehousing	-	28.2	11.8	31.2	11.4	27.9	10.8	24.5	12.3	28.8
(E) Length of service with employer										
, 3 months	12.6	-	12.1	-	10.4	-	7.8	-	8.5	-
3–11 months	20.5	-	19.0	-	21.1	-	16.8	-	15.2	-
1–5 years	30.9	-	32.4	-	36.3	-	37.6	-	35.7	-
- 5 years	35.6	-	35.9	-	32.0	-	37.2	-	39.5	-

Note: Slip or trip without fall data are not included.
Source: Bureau of Labor Statistics (2012).
[a] Listed in alphabetical order.

The percentage of fall-related injuries among selected occupational specialities is shown in Table 3(C). Overall, workers in the transportation and material-moving occupation had the highest amount of injuries associated with 'fall on same level' events that involved DAFW, with an average of 15.1% of total falls. In addition, workers in food preparation and serving, along with the office and administration support occupation, amassed the second and third highest percentage of injuries, with 10.5% of 'fall on same level' injuries. The office and administration support occupation comprises the following major groups: switchboard operators; bill and account collectors; clerks; cargo and freight agents; police, fire and ambulance dispatchers; postal service mail carriers; computer operators. With regard to incidence rates, healthcare support workers represented the highest rate of same level occupational fall-related injuries, equalling 41.7 falls in every 10,000 workers. The cleaning and maintenance occupation of the building and grounds was a close second with 40 falls, followed by the transportation and material-moving occupation with 30.5 falls.

Table 3(D) lists the types of industries frequently involved in 'fall on same level' injuries involving DAFW. The accommodation and food services industry has the highest number of falls with an average of 21.6% over the past 5 years, followed closely by the healthcare and social assistance industry with 20.1%; the retail trade industry with 15.4%. Figure 2 illustrates the incidence rate of the affected industries in 'fall on same level' related injuries. The workers in transportation and warehousing sector are at high risk of falls with an average of 28.1 falls in every 10,000 workers. Healthcare and social assistance workers are the second highest with an incidence rate of 27.9 falls per 10,000 workers. The industries such as accommodation and food services (incidence rate per 10,000 workers: 21.9); retail trade (19.1) and agriculture, forestry, fishing and hunting (18.9) experienced higher fall rates than the construction industry (15.3). Table 3(E) shows the length of service with the employer when the fall accident transpired. Workers exceeding a year of employment experienced _ 70% of fall accidents, whereas workers with under a year of employment endured considerably less (about 28.8%).

Table 4 summarises the consequences of occupational 'fall on same level' injuries involving DAFW. The lower extremities, which include the knees, feet and toes, were the most affected areas with an average of 30.7% of total fall- related injury categories. Likewise, the trunk, which encompasses both the shoulder and back, was the second most injured body part, with an average of 25.6%. The BLS uses the classification 'multiple body parts' for any injury in which body parts from two or more divisions of the body are injured. Workers with injured multiple body parts ranked third with approximately 21.8% of overall injuries (Table 4(A)). The BLS defines source of injury as 'the objects, substances, equipment, and other factors responsible for the injury or illness incurred by the worker or that precipitated the event or exposure' (US Department of Labor Bureau of Labor Statistics 2012). The sources of injuries associated with 'fall on same level' incidents are detailed in Table 4(B). Amongst these sources, the majority, i.e. 87.9%, resulted from floors, walkways or ground surfaces, while the sum of the remaining causal factors only culminated in 12.1% of injuries. Work absences as a direct result of injuries incurred on the job are presented in Table 4(C). The data show that nearly 30% of 'fall on same level' injuries resulted in a loss of 31 or more workdays

Table 5 summarises the compensated cost for the top 10 most serious workplace injuries and 'fall on same level' related injuries. The top 10 causes of workplace injuries includes overexertion, falls on same level, bodily reaction, falls to lower level, struck by object, repetitive motion, highway incident and others. The results show that over the 4-year period from 2006 to 2009, compensation costs grew 25% ($6.4 billion in 2006 to $7.9 billion in 2009) for the falls on the same level injuries.

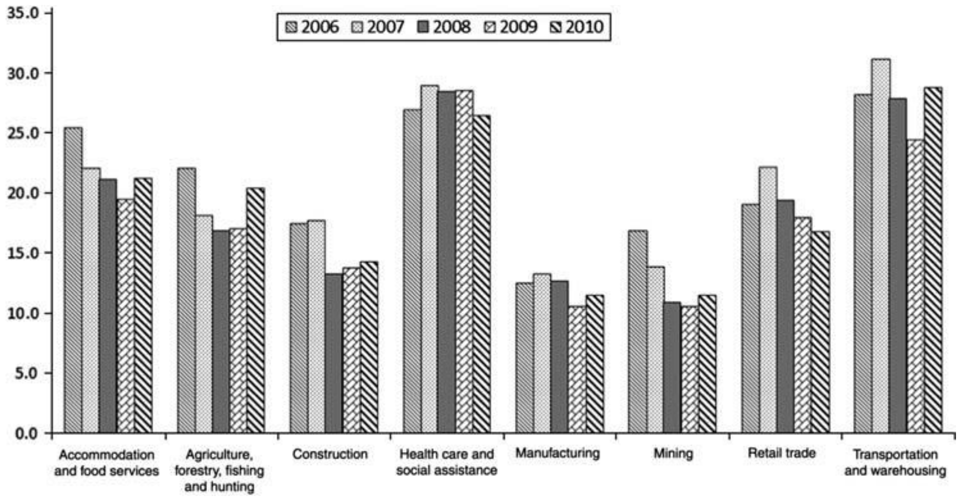

Figure 2. Number of incidences per 10,000 full-time workers for non-fatal occupational injuries in major US industries. Source: Bureau of Labor Statistics (2012).

Table 4. Consequences of same level falls involving DAFW in private industry, 2006–2010.

	2006		2007		2008		2009		2010	
Consequences of falls	Percentage	Rate	Percentage	Rate	Percentage	Rate	Percentage	Rate	Percentage	Rate
(A) Part of body injured										
Head	5.1	0.8	5.4	0.9	5.2	0.9	5.8	0.9	5.3	0.9
Neck	0.6	0.1	0.6	0.1	0.8	0.1	0.5	0.1	0.5	0.1
Trunk	25.8	4.2	26.1	4.6	26.2	4.3	25.7	4.0	24.0	3.9
Upper extremities	15.5	2.5	15.5	2.7	14.9	2.5	14.7	2.3	15.6	2.5
Lower extremities	31.2	5.1	30.7	5.4	29.5	4.9	30.5	4.7	31.8	5.1
Body systems	0.1	-	0.2	-	0.1	-	0.1	-	0.2	-
Multiple	20.7	3.4	21.1	3.7	22.7	3.8	22.2	3.4	22.2	3.6
(B) Source of injury										
Furniture, fixtures	2.4	0.4	2.9	0.5	2.6	0.4	2.8	0.4	2.4	0.4
Machinery	1.2	0.2	1.7	0.3	1.1	0.2	0.9	0.1	1	0.2
Parts and materials	1.9	0.3	1.4	0.2	1.3	0.2	0.9	0.1	1.1	0.2
Floor, ground surfaces	86.9	14.2	87	15.3	88.7	14.7	88.2	13.7	88.5	14.3
Vehicles	2.1	0.3	2	0.3	2	0.3	2	0.3	1.9	0.3
(C) Number of DAFW										
1 day	11.7	-	12.7	-	12.3	-	13.2	-	12.6	-
2 days	11.1	-	11	-	10.4	-	9.9	-	11.4	-
3–5 days	17.5	-	17.8	-	16.6	-	17.6	-	17.4	-
6–10 days	12.5	-	11.5	-	12.2	-	11.3	-	10.8	-
11–20 days	10.9	-	11.6	-	12	-	11.4	-	11.2	-
21–30 days	7.5	-	6.9	-	7.8	-	6.4	-	6.2	-
31 or more days	28.8	-	28.6	-	28.7	-	30.2	-	30.5	-

Note: Slip or trip without fall data are not included.
Source: Bureau of Labor Statistics (2012).

Table 5. Compensated cost for top 10 major occupational injuries and ‘fall on same level’ related injuries.

	All events	Fall on same level	
Year	Cost ($b)	Cost ($b)	Percentage
2006	48.6	6.4	13.2
2007	53.0	7.7	14.5
2008	53.4	8.4	15.7
2009	50.1	7.9	15.8

Note: Compensation costs for injuries involving six or DAFW only.
Source: Liberty Mutual (2010).

Discussion

Impact of 'fall on same level' occupational fall injuries

Although the number of overall occupational injuries involving DAFW decreased by 250,300 over the 5-year period (1,183,500 cases in 2006 to 933,200 cases in 2010), the percentage of overall fall injuries, including 'fall on same level', 'fall to lower level', 'jump to lower level' and other fall-related events, actually increased incrementally from 19.8% in 2006 to 22.3% in 2010. Furthermore, the percentage of 'fall on same level' injuries also increased in that manner from 12.8% in 2006 to 15% in 2010 (Table 1). The aforementioned decrease is likely a ramification of the high unemployment rate during this period as the unemployment rate in 2004 was a modest 5.5%, but rose to a staggering 9.3% in 2009 (BLS 2012). Thus, the injury percentage of 'fall on same level' accidents increased as it relates to overall occupational falls regardless of the unemployment rate. The results suggest that 'fall on same level' related occupational injuries contribute to ~ 14% of overall occupational injuries, and might be higher if we consider the number of 'slip, trip, loss of balance without fall' related injuries involving DAFW (around 3.2% of overall annual private industry injuries) which may lead to a fall incident. The finding on the incidence rate among fall events shows that 'fall on same level' events incur more injuries than 'fall to lower level' events by more than twofold.

Characteristics of injured workers

Gender. Female workers experienced a greater distribution and incidence rate of 'fall on same level' injuries involving DAFW compared to their male counterparts. The rationale behind this can vary, but the difference in the types of occupations between the two genders might be a contributing factor. Workers in food preparation and serving, along with the office and administration support, and healthcare support contributed to ~ 28% of 'fall on same level' related injuries, which are commonly female occupations. In addition, some believe that the task-related factors that are involved in load handling, patient lifting, rushing, risk-taking, physical exertion and complexity of task might increase the risks of fall (Gauchard et al. 2001; Lipscomb et al. 2006; Gao, Holmer, and Abeysekera 2008). Caregivers or nurses fell during patient transfers and when saving patients from a fall (Kemmlert and Lundholm 2001). Kemmlert and Lundholm (1998) also found that female healthcare workers had higher slips, trips and falls (STF) injury rates than male healthcare workers. Another factor might be pregnancy. Several studies have indicated that falls are a leading cause of maternal injury among pregnant women (Weiss, Sauber-Schatz, and Cook 2008; Schiff, Holt, and Daling 2002; Kuo et al. 2007). This is likely due to the musculoskeletal changes in a woman's body during pregnancy (i.e. increased joint and ligament laxity from hormonal changes) and shifts in the body's locus of balance from the increasing body weight and changing centre of gravity (Evenson et al. 2009; Ireland and Ott 2000; Butler et al. 2006). Furthermore, women in general are tasked with jobs that require formal wear and high heels as part of the dress code, i.e. in office and administration support occupation. With respect to shoe type, numerous safety organisations have cited high heels as a risk factor for slips and falls (Merrifield 1971; Snow and Williams 1994; Opila-Correia 1990). The correlation between high heels and increased potential for slipping suggests that the friction demand may be greater with high heels than with low heel shoes. For the purposes of this study, the results obtained here solely reflect the overall same level fall injuries related to gender and did not detail this parameter on specific industries. Previous investigations have shown that construction falls were the most prevalent for male workers, whereas most female worker falls occurred in the services

industry (Bunn, Slavova, and Bathke 2007). Therefore, the result highlights the need for more research to identify work-related exposures that influence the risk of fall in female workers.
Age. The labour force of the USA is growing older. This ageing is largely credited to the population growth during the baby boom era. The average retirement age is also expected to increase, due in part to a healthier older population, declining age discrimination, and gradual increases in age for collecting full social security benefits (Dohm 2000). These demographics suggest that issues of ageing will become more prominent in occupational safety and health research of the future (Layne and Pollack 2004). This is distressing as our results reported older workers have higher rates of STF involving DAFW accidents than younger workers, as has been reported elsewhere (Laflamme and Menckel 1995; Kemmlert and Lundholm 1998; Kemmlert and Lundholm 2001). Courtney et al. (2001) for instance, reported that the incidence rate corresponding with fall-induced fatalities increased sharply at 55 years of age and subsequently peaked in workers aged 65 and older. The consistency in these results and the aforementioned investigations suggests that fall accidents among the elderly may be related to age-related deterioration in the visual, proprioceptive and vestibular signals concerning postural control (Lockhart et al. 2002; Lockhart, Woldstad, and Smith 2003).

Occupation and industry. Slips and falls have been documented elsewhere as an important source of injury in jobs such as truckers and drivers (Nicholson and David 1985). The Miller study in 1976 indicated that about one-fourth of all truck driver injuries in the USA are associated with slips and falls in and around the truck (Miller 1976). While in 1997, the bureau of motor carrier safety, Federal Highway Administration, found that 54% of truck slip and fall accidents happen on the tractor and 46% on the trailer (Federal Highway Administration 1997). Workers in food preparation and serving amassed the second highest percentage of injuries, this occupation includes cooks, food preparation workers; bartenders, waiters and waitresses; food servers and dishwashers. The results presented here are consistent with those of the previous studies in which STF are one of the most common injuries in food services (Filiaggi and Courtney 2003; Courtney, Wellman, and Filiaggi 2005; Wellman, Filiaggi, and Courtney 2005; Alamgir et al. 2007; BLS 2007). Common sources of slippery floors in restaurant environment include dishwashing overspray or run-off, leaking equipment or pipes, food debris and spillage from transport of open containers (such as those holding fryer grease and food wastes) (Filiaggi and Courtney 2003). According to the Cotnam, Chang, and Courtney (2000), the healthcare industry is the largest employer in the USA (13 million employees) and ranks second among eight industries as having the highest percentage of claim costs associated with 'falls on the same level'. Some believe that this high percentage of falls is an indicative of their work environment, i.e. floor resistance, external environmental conditions and footwear (Veazie, Landen, and Bender 1994; Bell et al. 2008; Bentley and Haslam 1998). In addition, the nature of the work in these occupations or industries requires load carriage, and numerous studies have reported load carrying to be associated with an abnormal gait pattern and an increased heel slip distance after heel contact, hence increasing the risk of falls (Qu and Nussbaum 2009; Zultowski and Aruin 2008; Heller, Challis, and Sharkey 2009; Schiffman et al. 2006; Kincl et al. 2002; Park et al. 2010).
Length of service. The results show that workers exceeding a year of employment experienced ~70% of fall accidents involving DAFW, whereas workers with under a year of employment endured considerably less (28.8%). These findings suggest that employers could benefit by implementing an annual training programme to employees that reinforces safety awareness to prevent employees from future fall-related injuries. For example, employers are recommended under title 30 Code of Federal Regulations Part 46.8 to provide each miner with no 8 h of annual refresher training on fall prevention and protection (US Department of Labor 2011). Although the

incident rates are not available, studies elsewhere (Breslin and Smith 2006; Chau et al. 2010) have reported that workers with shorter job tenures have higher injury rates than workers with longer job tenures. For example, Chau et al. (2010) reported that the relative risk decreased steadily with increasing length of service within the company, from 2.6 for 1 year to 1.0 for $\geq$ 30 years in railway workers. Their findings suggest that direct experience in specific occupations is vital.

Consequences of occupational non-fatal 'fall on same level' related injuries

Part of body injured. Our study indicates that the majority of 'fall on same level' injuries involving DAFW were affiliated with the lower extremities. Similar results were obtained in other papers dealing with occupational ladder fall accidents (Cohen and Lin 1991; Cattledge et al. 1996). Although the fall accidents reported in this study are non-fatal injuries, it can still cause serious harm, such as bone fractures, back injuries, concussions or permanent disability (Courtney and Webster 2001; Courtney, Matz, and Webster 2002). The correlation between ageing and skeletal fractures, particularly female worker 'fall on same level' related injuries, cannot be underestimated as several studies have reported that women aged 45 years and above are at an increased risk of fracture due to falls (McNamee et al. 1997; Stevens and Sogolow 2005; Cherry et al. 2005).

Source of injury. Numerous countermeasures that could potentially reduce fall-related injury accidents include slip- resistant shoes, floor surfaces, mats, waxes, prompt cleaning of spillage and debris, keeping stairs and walkways clear, improving lighting, adding handrails and clearing ice and snow (Lewis 1997; Labar 1998; Morrison 1999). In 1985, Ballance et al. documented a reduction in the number of reported injury incidents involving 'falls on the same level' after replacing wood and ceramic flooring with less slippery tiles and carpet with higher coefficients of friction. Similarly, Manning et al. (1988) suggested that one of every four STF injury incidents could have been prevented by cleaning up spills and objects on the floor. Researchers have been working towards establishing safe floor resistance standards since the 1920s utilising the tribological approach (Biel 1920). Tribology deals with surface dissipative processes in terms of the hydrodynamics and viscoelastic characteristics of contaminants and the shoe-floor interface. The tribological approach to fall prevention has concentrated on setting safe static and dynamic coefficient of friction (COF) limits for ambulation (Chang et al. 2008; Redfern et al. 2001; Grönqvist et al. 2001). However, many conflicting ideas of dynamic versus static COF, lack of standard or law that requires that a floor must have certain level of slip resistance and ever-changing environmental conditions (associated with keeping the COF levels constant) hamper development in this area, and remain to be an important and critical task in the future.

Number of DAFW. Previous investigations from 1999 to 2001 only reported an average of 26.7% of injuries that culminated in 31 or more workday absences (Yoon and Lockhart 2005). This increase in DAFW suggests that the consequences of 'fall on same level' injuries are becoming more serious and that additional time is needed to recover. In addition, previous studies also suggest that older workers take longer time to return to work than their younger colleagues (Centers for Disease Control and Prevention 2011; Butler, Johnson, and Baldwin 1995; Rogers and Wiatrowski 2005), and that an increasing number of Americans have already begun delaying retirement to work longer. For example, between 1993 and 2009, labour force participation rates in the age group of 65-69 years increased from 25% to 36% for men and from 16% to 27% for women (BLS 2010b); furthermore, workers in the 55 years and older age group are projected to increase from 17% in 2006 to nearly 23% by 2016 (Toossi 2007; Garr

2009; Musich, McDonald, and Chapman 2009). We believe that with the increase in ageing workers in our society, fall-related injuries, especially 'fall on same level' injuries, need to be addressed.

Compensation cost of occupational falls on the same level

The National Safety Council reported that falls are the leading cause of death in the workplace and source of ~ 20% of all disabling injuries (National Safety Council 2002). Although the overall frequency of non-fatal 'fall on same level' events decreased, the proportion of occupational 'falls on the same level' injuries increased from 12.8% in 2006 to 15.0% in 2009 (2.2% during these 5-year period). Moreover, same level falls have consistently ranked second among the top 10 causes of serious workplace injuries in the USA (Liberty Mutual WSI 2000-2010). In addition, the costs of fall-related accidents are high since debilitating fall injuries such as back injuries increase the number of DAFW (National Occupational Research Agenda 1996). In addition, several studies claimed that using worker compensation claims to estimate the cost of injury underreports the actual burden of the injury; injured workers may choose not to file a claim for a variety of reasons: unfamiliarity about the system, legal status, fear of losing job or mistrust from filing with supervisors or even fellow employees (Azaroff, Levenstein, and Wegman 2002; Boden and Ozonoff 2008; Boden, Biddle, and Spieler 2001; Eisenberg and McDonald 1988). Boden, Biddle, and Spieler (2001) suggested that the true economic burden of an injury to an employer includes the indirect costs of hiring and training replacements, the impact on the productivity of new workers and coworkers and the administrative and supervisory time devoted to an injury. Other indirect costs include loss of teamwork and communication, and reduced motivation for training (Koopmanschap et al. 1995; Van Beek, Van Roijen, and Mackenbach 1997; Berger et al. 2001).

Limitation

Several limitations are noted. One of the main limitations is that the BLS data were subject to the employee's recollection of the incident and identification of the environmental risk factors. The narrative analysis method is limited by the completeness and consistency of the available text data (Lincoln et al. 2004). For instance, Lombardi et al. (2005) reported that it is not known whether words were truncated, forgotten, omitted or even lost in conversation by those reporting or recording the claim. There is growing evidence that the annual BLS SOII underestimates the true injury burden due to the underreporting of injuries (Azaroff, Levenstein, and Wegman 2002; Boden and Ozonoff 2008). A study by Leigh, Marcin, and Miller (2004) elaborated further, reporting that the Annual Survey missed from 33% to 69% of non-fatal injuries in 1999.

Another limitation is that the compensation costs taken from the Liberty Mutual (WSI) only reports the worker compensation cost for employees who missed 6 or more DAFW. Thus, we believe the actual number of falls and costs maybe higher than what is reported in this study. Although the extrinsic factors, such as floor surfaces, are reported by the BLS, the intrinsic factors such as past history of falls, medication usage, physical and mental conditions (tiredness, diseases and fatigue), which often increase the risk of falls, were not noted. Lack of information on confounding factors such as bodymass index is another limitation of this study, as obese adults fell almost twice as frequently (27%) as their lean counterparts (15%) per year (Fjeldstad et al. 2008), and are at a higher risk of fall (Matter et al. 2007; Wu, Lockhart, and Yeoh 2012). Moreover, this study was unable to determine the impact the organisation's safety culture may have had on fall injuries.

Despite the limitations, the findings allow us to form a useful picture around the risk and cost of 'fall on same level' injuries and have helped to determine where more information is required.

Conclusion

This paper describes the non-fatal 'fall on same level' injuries in a large population of private industry workers in the USA. Overall, workers in the healthcare and social assistance industry, the transportation and warehousing industry, and the accommodation and food services industry had the highest risk for non-fatal workplace 'fall on same level' injuries. Apropos to gender and age groups, female workers had the highest risk of falls, while advancing age was congruent with a rise in incidence rates. Over 30% of 'fall on same level' injuries resulted in 31 or more workdays being lost. This study also indicated that 70% of injured workers had 1 year experience with the company when the accident occurred. The most affected body parts were the lower extremities and the trunk. Floor and ground surfaces were the major determinants in fall injuries, and 'fall on same level' accidents have been ranked second as a leading cause of injuries during the 4-year period being studied.

This information can be used to prioritise designing and implementing preventive measures in US private industry and to provide workers with the understanding of risk factors associated with falls in the workplace. Leamon and Murphy (1995) concluded that 'based on the frequency and costs to industry and workers, prevention of falls should be given a high priority.' Preventive actions by the employer should be multidimensional, including a review of organisational practices and policies, work environment, health management programming and options for employees to update professional competencies. As the workforce ages, we recommend that educating older workers to be more aware of their physical and cognitive limitations may be the first step towards alleviating the impact of age on the risk of 'fall on same level' injury. In addition, employers can play a critical role by monitoring working arrangements and facilitating appropriate adjustments in the physical work environment. Further investigations of gender, age and occupation-specific prevention measures may be beneficial for workers.

Acknowledgements

Funding from the NSF (Grant#CBET-0756058) and NIOSH (Grant #CDC/NIOSH-R01-OH009222) supported this research. We acknowledge the help and cooperation of the Bureau of Labor Statistics and the National Safety Council, without which this study could not have been done. Our special thanks to Marcus Newland of the Bureau of Labor Statistics for his help in preparing the data.

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4. Aim 1: ***Effects of Fatigue on Slip Initiation***

The injury process associated with slip-induced falls includes three phases: 1) slip initiation, 2) slip detection, and 3) recovery. Slip initiation describes the personal, environmental and biomechanical conditions that dictate whether a given walking step will result in secure foot placement or if the foot will accelerate away from the base of support. Slip detection and recovery describes the neuromuscular and kinematic control sequence wherein the individual attempts to arrest the fall utilizing sensory mechanisms. Impact or fall occurs if the slip is initiated and recovery fails. Studies indicate that localized muscle fatigue can influence initiation, detection, and recovery phases of slip-induced falls.

The risk of slip initiation is directly related to the gait characteristics of the individual and the ground reaction force at the heel contact phase of the gait cycle (Lockhart et al., 2005). Initiation of a slip occurs when the frictional force (F_{μ}) opposing the movement of the foot is less than the horizontal shear force (F_h) at the foot during the heel contact phase of gait (Perkins, 1978, 1983). Specifically, at the time of heel contact, there is a forward thrust component of the foot against the floor resulting in a forward horizontal shear force (F_h). Additionally, a vertical force (F_v) occurs as body weight and the downward momentum loads the foot in contact against the ground. Frictional force is proportional to the vertical force, $F_{\mu} = \mu F_v$, with the constant of proportionality, μ, defined as the coefficient of friction. Hence, the coefficient of friction $\mu = F_{\mu}/F_v$ of the foot-ground interaction must be greater than the ratio F_h/F_v to avoid slip initiation. Perkins (1978) identified six peak forces in a normal gait cycle by measuring ground reaction forces exerted between the shoe and ground on a non-slippery floor surface. The ratio of horizontal to vertical foot forces (F_h/F_v) was also calculated. The significance of this ratio (F_h/F_v) is that it indicates where in the walking step a slip is most likely to occur (slip initiation). Analyzing the ratio, it was noted that a dangerous forward slip was most likely to occur within 50-100 ms of the heel contact phase of the gait cycle. This ratio (F_h/F_v) was called "Required Coefficient of Friction" (RCOF) (Redfern and Andres, 1984; Gronqvist et al., 1989), because it represented the general friction demand, i.e., minimum coefficient of friction that must be available or "required" at the shoe-floor interface to prevent initiation of forward slipping. The number of slip and fall events increases as the difference between the RCOF and available dynamic COF of the floor surface increases (Redfern et al., 1997). Thus, changes in RCOF as a function of localized muscle fatigue can provide insight into the risk of slip initiation.

Localized muscle fatigue of the lower extremities induced by prolonged walking influences initial gait characteristics such as step length, heel contact velocity, and ground reaction forces. Saggini et al., (1998) examined the effects of localized muscle fatigue on the lower extremity and concluded that fatigue increased the gait cycle time and also increased the heel contact velocity. Initial gait characteristics such as heel contact velocity affect RCOF by altering the ratio of horizontal to vertical foot forces (Lockhart et al., 2003). Increases in horizontal heel velocity during a critical time of weight transfer may increase the potential for slip-induced falls if the friction between the heel and the floor is reduced due to contamination of the floor surface. For example, investigations of older individuals' gait characteristics by Lockhart et al. (2000) and Winter (1990) revealed that the risk of slip-induced falls was higher due to the higher heel contact velocity. A likely factor influencing the higher horizontal heel contact velocity may be a decrease in hamstring activation rate due to localized muscle fatigue. These muscles become active at the termination of swing phase, being elongated as they act to decelerate the swing leg, and help extend (control) the knee and hip. As discussed earlier localized muscle fatigue is defined as an acute impairment in the ability to exert force or power, and fatigue occurs as the metabolic by-products reduce intra-muscular conduction velocity until the muscles become unable to produce the desired forces

(Svantesson et al., 1998; Gefen et al., 2002). Fatigue-induced contractile process and excitation-contraction coupling failure may decrease the hamstring activation leading to higher heel contact velocity. Thus, effects of localized muscle fatigue on gait characteristics and specifically higher horizontal heel contact velocity during the critical phase of the gait cycle, may increase the friction demand (i.e., RCOF) and, thereby, increase the likelihood of slip-induced fall accidents.

Furthermore, the onset of lower extremity fatigue during walking changed the loading rate and increased ground reaction forces (Syed and Davis, 2000) thereby reducing the forward momentum of the whole-body COM. Lockhart et al., (2003) indicated that reduced push-off force of the stance leg further reduced the transitional acceleration of the whole-body COM and increased RCOF and the risk of slip initiation. In other words, a reduction in the transitional acceleration of the whole-body COM due to localized muscle fatigue is likely to increase the friction demand at the shoe/floor interface of the contacting foot. Increased initial friction demand (i.e., RCOF) would lead to a higher likelihood of slips associated with low coefficient of friction floor surfaces. A likely factor influencing the transitional acceleration of the whole-body COM may be the ankle plantar flexors' biomechanical and physiological factors – i.e., plantar flexors produce more than half of the positive work during the push-of phase of the gait cycle (Winter, 1983). Additionally, smooth transition of the whole-body COM is maintained by hip flexion/extension to reduce the jarring effects (Inman et al., 1981). In summary, alterations of the ground reaction forces and gait kinematics (heel contact velocity and transitional acceleration of the whole-body COM) due to localized muscle fatigue of the lower extremities may increase RCOF and risk of slip initiation. In this section, two studies are presented to represent different fatigue locations. The lower extremities fatigue was induced at quadriceps and gastrocnemius muscle groups. The quadriceps fatigue experiment was performed during the period of the review of the proposal in 2008 and thus it is included in this report. Additionally, ankle fatigue experiment was conducted during 2009-2010 period and is currently being submitted to a journal publication. Overall, fatigue influenced slip initiating characteristics and differential effects were observed between different fatigue locations.

4A. Aim 1: ***<u>Effects of Fatigue on Slip Initiation: Gait Changes</u>***

Zhang, J., Lockhart, T.E., and Soangra, R., (2014), Classifying Lower Extremity Muscle Fatigue during Walking using Machine Learning and Inertial Sensors. *Annals of Biomedical Engineering, 42(3): 600-612.*

Abstract

Fatigue in lower extremity musculature is associated with decline in postural stability, motor performance and, alters normal walking patterns in human subjects. Automated recognition of lower extremity muscle fatigue condition may be advantageous in early detection of fall and injury risks. Supervised machine learning methods such as Support Vector Machines (SVM) have been previously used for classifying healthy and pathological gait patterns and, separating old and young gait patterns. In this study we explore the classification potential of SVM in recognition of gait patterns utilizing an inertial measurement unit (IMU) associated with lower extremity muscular fatigue.

Both kinematic and kinetic gait patterns of 17 participants (29±11 years) were recorded and analyzed in normal and fatigued state of walking. Lower extremities were fatigued by performing a squatting exercise until the participants reached 60% of their baseline maximal voluntary exertion level (most of the participants reached this state within 45-60 minutes of isotonic exertions). Three different feature selection methods were used to classify fatigue and no-fatigue conditions: (1) selected "ad hoc" features based on domain knowledge; (2) general features; and (3) complete waveforms of kinematic data from the IMU situated at the trunk Center-of-Mass (COM) location. Additionally, influences of three different kernel schemes used in SVM classification method were investigated.

The results indicated that lower extremity muscle fatigue condition influenced gait and loading responses (i.e., jerk cost). In terms of the SVM classification results, an accuracy of 96% was reached in distinguishing the two gait patterns (fatigue and no-fatigue) within the same subject using the kinematic time and frequency domain features. It was also found that linear kernel and radial basis function kernel were equally good to identify intra-individual fatigue characteristics. These results suggest that intra-subject fatigue classification using gait patterns from an IMU holds considerable potential to identify at-risk gait due to muscle fatigue. Additionally, the paired t-test results indicated that participants increased their step width ($p<0.02$), reduced the single stance duration ($p<0.01$), and increased the jerk cost ($p<0.03$) in post fatigue walking trials. It was also found that fatigue significantly increased slip propensity as measured by the heel contact velocity ($p<0.01$). The SVM method also classified inter-subject fatigue condition with an accuracy of 75% and 90% using the forceplate and an IMU gait patterns respectively. In summary, kinematics of the trunk COM may provide additional discriminatory information for improved fatigue state classification performances using support vector machines.

Key terms: Locomotion, Machine Learning, Support vector machines, Jerk Cost, fall risk, falls, classification

Introduction

The automated recognition of muscle fatigue through gait patterns may be important because of its potential diagnostic applications in various working environments. For example, early detection of locomotor impairments would provide the opportunity to identify at-risk gait due to muscle fatigue and, administer corrective measures such as rest- breaks or change in worker pool. In this study we explore the classification potential of SVM in recognizing gait patterns associated with lower extremity muscular fatigue utilizing an inertial measurement unit (IMU) as the wearable technology has the potential to investigate continuous kinematic changes evoked by fatigue [1]. “Fatigue” is a loosely defined term and is a symptom associated with various chronic diseases [2]. Self-reported fatigue implies a feeling of exhaustion, lethargy, reduced activity and, strength. Muscle fatigue is a time-dependent feature influencing the maximal force generating capacity of a muscle. It is also considered as a positive phenomenon, which protects muscle tissue from damage that may incur due to overuse. Localized muscle fatigue is a potential risk factor for slip-induced falls [3] as muscle fatigue adversely affects proprioception [4-6], movement coordination and muscle reaction times [7] leading to postural instability [8] and gait changes [3, 9]. Although human locomotion is programmed in the central nervous system (CNS), it is adapted by proprioceptive feedback [10]. During stance phase of the gait cycle, proprioceptive input from extensor muscles and mechanoreceptors in the sole of foot provide load information [11] to the CNS. As the afferents that signal hip joint position coming mainly from muscles around hip contribute to activation pattern of lower extremity muscles, fatigue in lower extremity muscles may increase irregular firing of motor neurons during walking [12-14]. As such, fatiguing of the muscles around a joint inhibits the joint’s neuromuscular feedback and influence synergism between joint proprioception and muscular function leading to instability and gait changes [15-23].

In our previous work, we observed that localized muscle fatigue of the quadriceps affected various kinematic and kinetic gait parameters that are linked with a higher risk of slip-induced falls in healthy young adults [3]. A delayed response to a slip perturbation was also observed in producing counter reactive joint moment in fatigued state. Fatigue induced by resistance exercises to lower limbs and trunk muscles has been shown to induce postural instability and impairs functional reach, reduces speed and power of sit-to-stand and, found to produce less stable and more variable walking patterns [24].

Limited information exist in understanding the impact of muscle fatigue on dynamic postural control during walking that may be amendable to classification schemes [25] [9]. Human locomotion involves motor control of lower extremities utilizing visual, vestibular, proprioceptive feedback and, neuromuscular responses. Recent studies assessing the effect of muscle fatigue on balance control during walking indicated compensatory changes in trunk and head movements following fatigue [9, 26]. Additionally, older adults took significantly wider steps with greater step variability and, trunk acceleration variability during post fatigue walking trials [10]. Furthermore, one of the pioneering works investigating the effects of muscle fatigue during running indicated that most promising sensor locations were the trunk and the foot [1]. The most important kinematic parameters that changed during fatigue were: (1) vertical oscillations; (2) impact acceleration on upper body; (3) trunk forward leaning; and (4) shoulder rotation. This information may be used to classify fatigue states utilizing SVM.

Numerous classification algorithms exist to provide human motion classification patterns. Najafi et. al. used gyroscope data and wavelet method to analyze the “sit-to-stand” transition in relation to the fall risk [27]. Lee et al. proposed linear discriminant analysis method to classify external load conditions during walking [28]. Begg et. al. used the SVM classifier to analyze the minimum foot clearance owing to aging [29]. The SVM is considered a powerful technique for general data classification and has been widely used to classify human motion patterns with good results [30-33]. The advantage of SVM algorithm is that it can generate a classification result with limited data sets by minimizing both structural and empirical risks[34]. Although numerous studies have been devoted to improving the SVM algorithms, little work has been performed to assess the robustness of SVM algorithms associated with movement variations and fatigue states. Furthermore, existing analysis are mainly based on motion capture systems and force plate measurements. While these systems are highly accurate, they do not allow continuous monitoring outside laboratory environments. Additionally, the cost of commercially available motion analyses systems and complexity of data analysis of such systems restrict their use to research environments only.

In the current study, we aim to monitor kinematics of walking in unconstrained environments using an IMU situated around the trunk COM during fatigue and no-fatigue walking conditions. We hypothesize that lower extremity muscle fatigue will influence walking behavior and this subtle changes in gait can be classified by supervised machine learning techniques such as support vector machines.

Materials and Methods

Participants: Seventeen healthy young adults (9 males and 8 females) participated in the study. The participants mean age was 29±11 years, height 174±10 cm, and weight 73±12 kg. Testing lasted 3 hours. In this total time, fatigue procedure took about 45-60 minutes. The experiment was composed of inducement of fatigue in lower extremity joints (ankle, knee and hip) with squatting exercises [35, 36]. Walking trials were conducted both prior and after the fatiguing condition. All participants were healthy, independent and non-sedentary and, were formally screened for major musculoskeletal, cardiovascular, and neurological disorders by a research coordinator during initial participant contact. Exclusion criteria of this study were factors which could interfere with gait, such as medication use, presence of neuromuscular disease and, balance and vision disorders. Participants were instructed not to perform any strenuous exercise 24h before the experiment. All experiments were conducted between 11:00 AM and 4:00 PM, and this was conducted to control the circadian effects of fatigue. Informed consent was approved by the Institutional Review Board (IRB) of Virginia Tech and was signed by all participants prior to the study.

Instrumentation: The IMU node consisted of MMA7261QT tri-axial accelerometers and IDG-300 (*x* and *y* plane gyroscope) and ADXRS300, *z*-plane uniaxial gyroscope aggregated in the Technology-Enabled Medical Precision Observation (TEMPO) platform which was manufactured in collaboration with the research team of the University of Virginia [37]. The data acquisition was carried out using a Bluetooth adapter and Laptop through a custom built LabView VI. Data was acquired with sampling frequency of 128Hz. This frequency is largely sufficient for human movement analysis in daily activities, which occurs, in low bandwidth [0.8-5Hz] [38]. The data was processed using custom software written in Matlab (the Mathworks, Inc.) and libSVM toolbox [39].

Walking trials were conducted on a linear walkway (1.5 X 15.5 m) embedded with two force plates (Bertec Corporation, Columbus, OH, USA). A total of 23 reflective markers were placed over various bony landmarks of the participants. The marker configuration of the whole body model was similar to previous study by Parijat & Lockhart [3]. Two IMUs were affixed on the participants, one at right shank (to normalize the gait cycle) and the other at the sternum level. Walking trials were preceded by laboratory environment acclimatization and warm-up period of 10 minutes. Participants were asked to walk at their self-preferred pace on the walkway and kinematic, kinetic, inertial data was collected. Gait characteristics were assessed in the middle portion (5 m) of the walkway. Ground reaction forces were measured (sampling frequency 1200Hz) from the force plates positioned across the center area of the walkway.

Fatigue Inducement Protocol: A custom built Biodex attachment for the shoulders was used to assess maximum voluntary isokinetic exertions (MVE) during squatting. The Biodex attachment was designed to measure combined torque from the ankles, knees and, hips through vertical motion/force exerted via shoulders. Although MVE trials were performed using the shoulder attachment with a dynamometer during squat protocol, fatigue was induced with carrying 5% body weight while squatting repeatedly at 22 repetitions per minute [35]. All exercises were performed at regular intervals (i.e., after 5min exercise, MVE was measured) until the participants reached 60% of their baseline MVE; this was categorized as fatigued state (most of the participants reached this state within 45-60 minutes of isotonic exertions).

Each session consisted of both normal walking and post-fatigue walking trials. Ground reaction force data was reviewed in every trial for ascertaining foot placement in the desired sequence (i.e., left-right heel contacts). Three trials of gait cycles were recorded using the forceplates. In all, ten gait cycles were collected for kinematic analysis by the IMU. Normal walking trials were preceded by a warm-up phase (walking back and forth three times around the laboratory). In post-fatigue walking trials, participants walked back and forth at their preferred speed. And no warm-up was performed after isotonic fatiguing exercises. Timeline of testing procedure is illustrated in Figure 1.

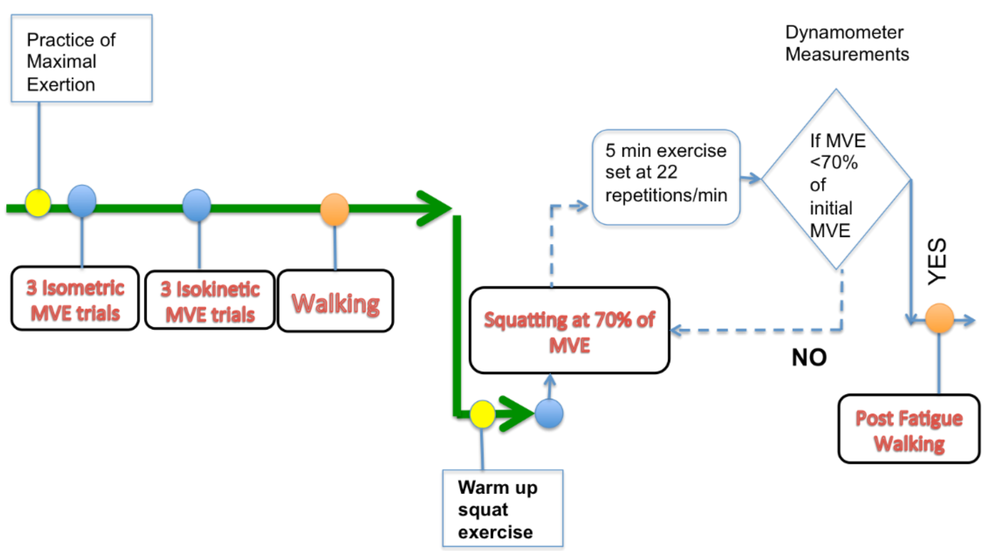

Figure 1: Timeline of testing procedures. Practice trials were performed with the dynamometer followed by data collection for maximum voluntary torque and normal walking. Before inducing fatigue warm up exercises were conducted. Participants were considered fatigued when their isokinetic MVC was below 70% of initial isokinetic values for all three trials.

Data Analysis: For the classification, both training and testing data sets consisted of fatigue and no fatigue walking data. Kinematic data used for SVM input was Representative Gait Cycle (RGC) data. RGC begins when one foot contacts the ground and ends when that foot contacts the ground again using the shank IMU. A perfect representative gait cycle signal between two easily identifiable events of the same foot was chosen for the analysis (Figure 2). This representative gait cycle started at peak right shank angular velocity (left foot mid-stance) and terminated at consecutive peak right shank angular velocity (left foot mid-stance). All IMU signals were truncated to RGC and normalized to 0% to 100%. A repeated-measure design was used to test changes within-subject in gait parameters from normal walking and post fatigue walking trials. A paired sample t-test was used to test the parameters.

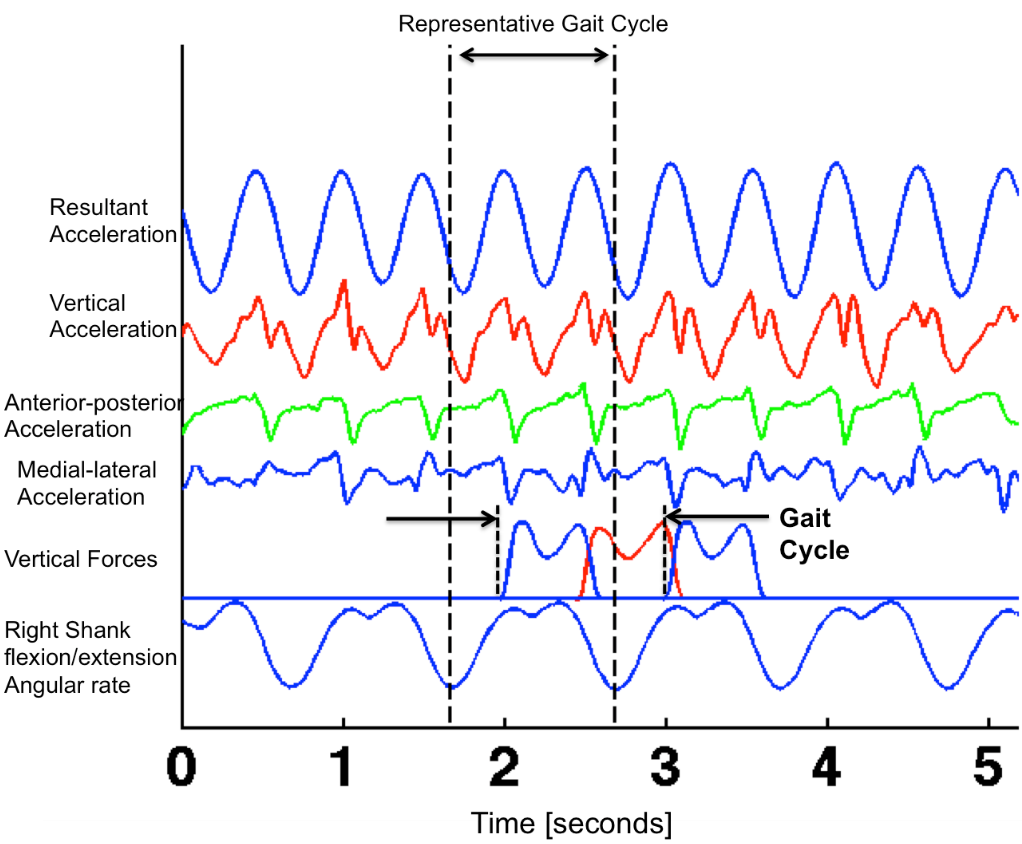

Figure 2: Two consecutive time epochs when right shank attains peak angular velocities were chosen during walking as input gait pattern data mimicking gait cycle and was defined as Representative Gait Cycle. The R-GC data from IMU situated at trunk was truncated for extraction of features values to SVM.

Data Input to the SVM Classifier: Support Vector Machines (SVM) is a powerful technique for general data classification and has been widely used to classify human motion patterns with good results [31-33]. The advantage of SVM algorithm is that it has exceptionally good performance for binary classification tasks [30, 40-43] and can generate a classification result with limited data sets by minimizing both structural and empirical risks [34]. SVM maps the original data to a high-dimensional space using a nonlinear mapping by finding an optimum linear separating hyper-plane with the maximal margin in this higher dimensional space [44, 45] utilizing gait features. The purpose of feature extraction is to retain as much of spatial and temporal information of the gait cycle as possible. Complete description of SVM is attached in the Appendix A. The SVM classifier has not been applied previously to fatigue and no-fatigue gait patterns. An important characteristic of using SVM classifier in this study was to obtain high fatigue/no-fatigue classification accuracy with three different types of feature input (Table 1): (1) selected “ad hoc” features based on domain knowledge; (2) general features; and (3) concatenated complete gait pattern signals. Figure 3 summarizes the procedure followed for SVM classification.

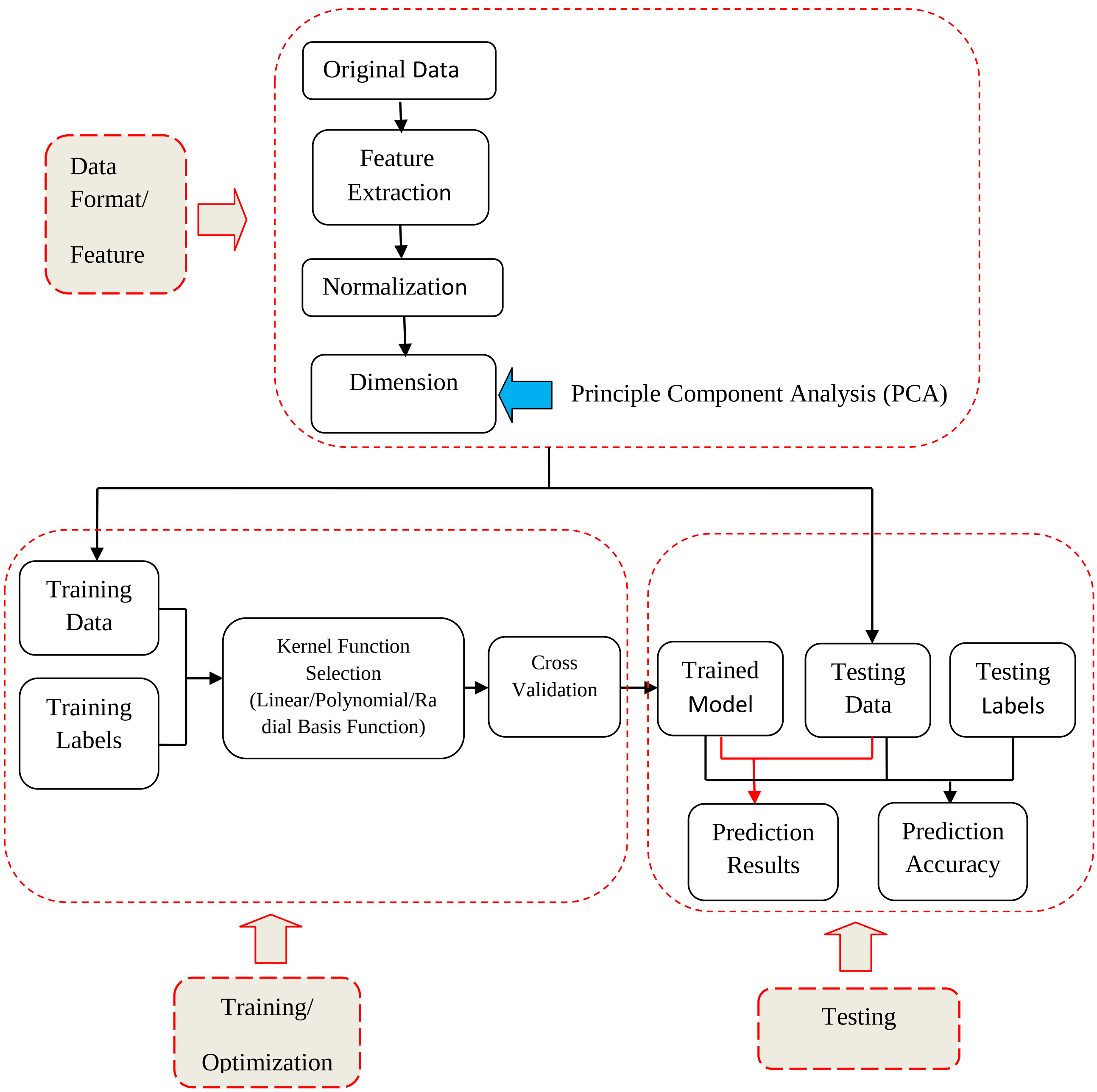

Figure 3: Schematic diagram of procedure of SVM classification.

Selected Features: In total, 11 kinematic features were selected from resultant walking COM acceleration and jerk. Jerk cost, as described by the area under squared jerk curve is an important measure to estimate the energy economy of walking. During walking, minimizing jerk and minimizing energy are believed to be complementary performance criteria [46, 47]. Resultant acceleration of the trunk COM and derived features such as maximum, minimum, range, energy and dominant frequency while walking are important as they provide complete kinematics of the trunk COM. Helbostad and his colleagues have reported significant increase in trunk acceleration due to physical fatigue [9]. Skewness of resultant acceleration provides information of the temporal shift of peak accelerations in RGC. Figure 4 illustrates resultant acceleration and jerk profiles for a complete RGC. We performed the classification with possible kinematic features, which could bring significant changes due to fatigue (Table 1). The kinetic features such as peak forces during heel-strike, mid-stance and push-off phases were extracted and normalized from heel contact (0%) to toe off (100%). All together 5 features were selected describing kinetic aspects of gait data [31].

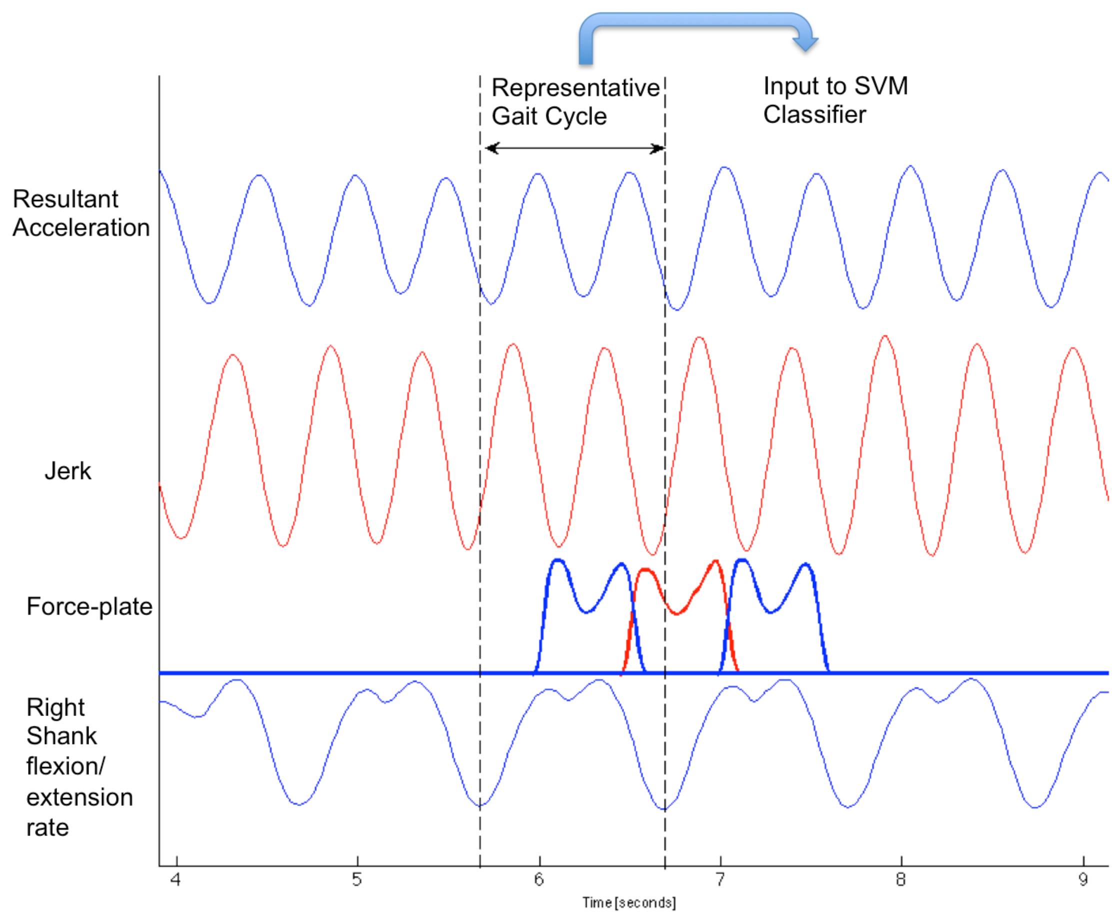

Figure 4: Resultant acceleration and jerk of COM measured through IMU was truncated to R-GC data and used to evaluate selected feature input for SVM.

General Features: The general features were chosen to include all possible spatial and temporal information from the signals. Based on the criterion of minimizing computational complexity and maximizing the class discrimination, several key features are proposed for SVM classification [48].

Mean Absolute Value - the mean absolute value of the original signal, $\bar{x}$, in order to estimate signal information in time domain:

$$\bar{x} = \frac{1}{N}\sum_{k=1}^{N}|x_k| \qquad (1),$$

where x_k is the k_{th} sampled point and *N* represents the total sampled number over the entire signal.
Zero Crossings - the number of times the waveform crosses zero, in order to reflect signal information in frequency domain.

Table 1: Three feature sets were used as inputs to SVM. 1) General features, 2) Selected features and 3) Complete concatenated waveform.

	General Features		**Domain knowledge based Selected Features**		**Complete Concatenated Waveform**	
	Kinematic input	**Kinetic input**	**Kinematic input**	**Kinetic input**	**Kinematic input**	**Kinetic input**
Data input for SVM	Accelerometer and gyroscope data in all 3 directions of normalized representative gait cycle	Forces and moments in all 3 directions of normalized stance data	-Accelerometer and gyroscope data in all 3 directions of normalized representative gait cycle - Resultant acceleration -Resultant Jerk	Vertical and medial-lateral forces	Concatenated input of normalized representative gait cycle	Concatenated input of forces and moments in all 3 directions of normalized stance data
	i) mean ii) standard deviation iii) maximum iv) minimum v) Mean Absolute Value $\bar{x} = \frac{1}{N}\sum_{k=1}^{N}\|x_k\|$	i) Mean ii) standard deviation iii) maximum iv) minimum v) Mean Absolute Value	Resultant Acceleration features i) skewness (temporal shift) ii) Energy iii) Dominant frequency iv) Maximum acceleration v) Minimum Acceleration Range of acceleration	i) Maximum vertical Heel contact Force ii) Vertical min mid-stance force iii) Max vertical push-off force	--No Feature--	--No Feature--
	i) skewness ii) kurtosis iii) Energy iv) Number of Slope sign changes v) Number of zero crossings	vi) skewness vii) kurtosis viii) Energy ix) Number of Slope sign changes x) Number of zero crossings	Resultant Jerk features i) Skewness (temporal shift) ii) Mean Jerk at heel contact iii) Absolute Maximum Jerk iv) Absolute Minimum Jerk	iv) Max horizontal Heel contact force during braking phase v) Max horizontal push-off force		
	xi) Length of waveform xii) Dominant Frequency using low-pass filter and FFT	xi) Length of waveform xii) Dominant Frequency using low-pass filter and FFT	v) Range of Jerk Produced abs(max-min) vi) Jerk Cost $JC = \int_0^T \|\frac{d^3r}{dt^3}\|^2 dt$			

Slope Sign Changes - the number of times the slope of the waveform changes sign, in order to measure frequency content of signal.
Waveform Length - the cumulative curve length over the entire signal, in order to provide information on the waveform complexity. All of these features would give a measure of waveform amplitude, frequency, and duration within a single parameter. Table 1 elaborates general features used in this study.

Complete Concatenated Waveform: All RGC signals from the IMU were concatenated for input to SVM classifier. Similarly all forces and moments across all three directions were used as concatenated force plate data input to the classifier.

Data Processing:
A) *Normalizing Input Data*: All features or input variables were normalized by computing their z-scores. Input data was kept in range between 0 and 1.

B) *Dimension Reduction of Feature Space*: Principle Component Analysis (PCA) [49] was employed to decrease the dimensions. The objective of PCA is to perform dimensionality reduction while preserving as much of the randomness in the high-dimensional space as possible. The PCA algorithm can be described as:
1) Assume $\boldsymbol{X}$ is an $m \times n$ matrix, and we choose a normalized direction in m-dimensional space along which the variance in $\boldsymbol{X}$ is maximized, saving this vector as $\boldsymbol{p_1}$.
2) Find another direction along which variance is maximized, however, the search is restricted to all directions orthogonal to all previous selected directions, due to the orthonormality condition, saving this vector as $\boldsymbol{p_i}$. The procedure is repeated until m vectors are selected. The resulting ordered set of $\boldsymbol{p}$'s are called principal components.

The dimension reduction can be described as:
1) For the m eigenvectors, we reduce from m dimensions to k dimensions by choosing k eigenvectors related to k largest eigenvalues λ;
2) Proportion of Variance (PoV) can be explained as:

$$\mathrm{PoV} = \frac{\lambda_1+\lambda_2+\cdots+\lambda_k}{\lambda_1+\lambda_2+\cdots+\lambda_k+\cdots+\lambda_m} \quad (2),$$

where λ_i are sorted in descending order, and the threshold of PoV is typically set as 0.9.

C) *Cross-validation*: A five-fold cross-validation scheme was adopted to evaluate the generalizability of the SVM classifier [31, 50]. In cross-validation procedure, the training data set is uniformly divided into five subsets with one used for testing and the other four used for training and constructing the SVM decision surface. This process is continued until all subsets are used as the testing sample.

D) *SVM Classifier Testing*: All SVM models were trained over the range C=2^{-10} to 2^{10} using linear, polynomial and radial basis function kernel. Three criterions were used to assess the performance of SVM classifier in classification.

$$Accuracy = \frac{TP+TN}{TP+FP+TN+FN} \times 100\% \quad (3)$$

$$Sensitivity = \frac{TP}{TP+FN} \times 100\% \quad (4)$$

$$Specificity = \frac{TN}{TN+FP} \times 100\% \quad (5),$$

where *TP* represents the number of true positive, SVM identifies a normal no-fatigued gait that was labeled as no-fatigue; *TN* is the number of true negatives, SVM identifies fatigued gait data that was labeled as fatigue; *FP* is the number of false positives, false no-fatigue identification; and *FN* is the number of false negatives, false fatigue identification. While accuracy indicates overall detection accuracy; sensitivity is defined as the ability of the SVM classifier to accurately recognize no-fatigue condition; and specificity would indicate the SVM classifier's ability to avoid false detection. Schematic diagram of SVM classification algorithm is illustrated in Figure 3.

Furthermore, Receiver Operating Characteristic (ROC) curve was also used to evaluate SVM classifier's performance. ROC analysis is generally utilized to select optimal models and to quantify the accuracy of diagnostic tests. Besides, the Area Under the ROC Curve (AUC), which is a representation of the classification performance, was utilized to assess the effectiveness of SVM classifier. Tests were also conducted to evaluate performance of the SVMs for three different kernel functions. Linear, Polynomial and Radial Basis Function (RBF) kernels were employed for classification.

Results

Table 2 shows the comparison between gait parameters with inducement of fatigue. No significant differences were found for the step length of the participants due to fatigue. However, it was seen that participants adopted wider base of support in post fatigue walking trials. Fatigue increased step width by about 12% as compared to the no fatigue walking. Although no statistical significant ($p>0.05$) difference was observed in walking velocity, heel contact velocity was increased ($p=0.01$) in post-fatigue walking trials.

Table 2: Step length (mm), step width (mm), heel contact velocity (mm/sec), walking velocity (m/sec) and stance duration (seconds) were evaluated for no fatigue and post fatigue walking trials. The data provided is in mean ± SD for the group and paired t-test was used with alpha set at 0.05.

	No Fatigue	**Post Fatigue**	**p-Value**
Step Length (mm)	755±59	748±58	>0.05
Step Width (mm)	110±32	123±38	0.02*
Heel Contact Velocity (mm/sec)	569±110	637±122	0.01*
Walking Velocity (m/sec)	1.41±0.15	1.40±0.16	>0.05
Single Stance Time (sec)	0.68±0.04	0.66±0.04	<0.01*

The machine learning classification results demonstrated high intra-individual classification rates across all three-kernel types (i.e., linear, polynomial and radial basis function kernel). We found that linear (accuracy ~96-97%) and RBF (accuracy 96%) kernels perform equally well in intra-individual fatigue/no-fatigue classifications (Table 3). Normalized complete concatenated waveform input signals for SVM classification was accurate in distinguishing classification for intra-individual gait signals however, failed to distinguish when used for inter-subject classification. The polynomial kernel had the lowest classification accuracy amongst all three different types of kernels.

Table 4 shows mean success rates of SVM classifier for inter-subject fatigue classification. In this table, accuracies have been compared for both forceplate and the IMU derived feature input. The average success rate was about 75% using the forceplate information alone (employing general and selected feature input to SVM). It is important to note that poor inter-subject fatigue classification accuracies (~62%) were obtained when normalized complete concatenated waveform (forces and moments) was used as input to SVM. RBF kernels provided the best separation accuracies for fatigue and no-fatigue walking trials with higher sensitivities and specificities relative to the linear and polynomial kernels.

Table 3: Intra-subject fatigue classification using IMU derived features. Accuracy, sensitivity, specificity and AUC (area under the Receiver operating curve) are tabulated for three kinds of feature selections methods and three kernels.

	Combined Joint Fatigue			
		Linear	Polynomial	RBF
General Features	Accuracy	97	88	96
	Sensitivity	98.03	92.15	98.03
	Specificity	96.07	84.31	94.11
	AUC	0.98	0.98	0.98
Domain knowledge based Selected Features	Accuracy	93	86	93
	Sensitivity	90.19	82.35	88.23
	Specificity	96.07	90.19	98.03
	AUC	0.96	0.94	0.97
Normalized complete concatenated waveform signals	Accuracy	96	80	96
	Sensitivity	96.07	66.66	98.03
	Specificity	96.07	94.11	94.11
	AUC	0.97	0.97	0.98

SVM achieved about 90% inter-subject fatigue classification accuracies with IMU general features for identifying fatigue among participants (Table 4). Selected features from the trunk COM kinematics could achieve a good accuracy of 88%. Additionally, these selected set of features were analyzed statistically for both fatigue and no-fatigue conditions. We found that features of resultant acceleration and jerk such as maximum; minimum, range, skewness, and energy along with jerk cost were significantly different for post-fatigue walking as reported in Table 5 . Dominant frequency was not significantly different for no-fatigue and post-fatigue walking conditions. It was seen that accuracy increased from 87% to 88% by inclusion of this feature in the selected feature set.

Table 4: Inter-subject fatigue classification using forceplate and IMU derived features. Accuracy, sensitivity, specificity and AUC (area under the Receiver operating curve) are tabulated for three kinds of feature selections methods and three kernels.

		Force Plate			Inertial Measurement Unit		
		Linear	Polynomial	RBF	Linear	Polynomial	RBF
General Features	Accuracy	75	75	75	88	90	90
	Sensitivity	60	60	60	88	92	92
	Specificity	81	81	81	88	88	88
	AUC	0.76	0.65	0.80	0.93	0.92	0.95
Domain knowledge based Selected Features	Accuracy	68	62	75	85	85	88
	Sensitivity	0	100	40	80	82	86
	Specificity	100	45	90.2	90	88	90
	AUC	0.54	0.92	0.63	0.93	0.92	0.94
Normalized concatenated signals	Accuracy	62	50	62	71	84	90
	Sensitivity	40	40	40	63	73	88
	Specificity	72	54	54	80	96	92
	AUC	0.61	0.49	0.65	0.73	0.90	0.92

Table 5: Selected features from IMU were computed for no-fatigue and post fatigue walking and statistical analysis is reported.

Gait characteristics from IMU	Features	No-Fatigue Walking	Post-Fatigue Walking	p-Value
Resultant Acceleration	Maximum [g]	0.20	0.42	<0.01
	Minimum[g]	0.04	0.12	<0.01
	Range[g]	0.15	0.30	<0.01
	Skewness	0.21	0.07	<0.03
	Energy [g^2. sec]	2.14	10.15	<0.01
	Dominant Frequency [Hz]	1.4	1.5	=0.15
Jerk	Maximum[g/sec]	0.01	0.02	<0.01
	Minimum[g/sec]	0.009	0.01	<0.01
	Range[g/sec]	0.02	0.04	<0.01
	Skewness	0.18	0.31	<0.05
	Jerk at Heel Contact[g/sec]	0.003	0.007	<0.01
	Jerk Cost [g^2/sec]	0.007	0.02	<0.03

It was observed that for inter-subject fatigue classification with training sets of 238 and testing set of 102 data sets (i.e., 17 participants had 20 set of data -10 normal walking, and 10 fatigue walking), linear kernel required 70.85 seconds, polynomial kernel required 15.18 seconds and radial basis function required 16.59 seconds for training the model on 2.2GHz Intel Core i7 processor and Matlab R2012b. The inter-subject fatigue classification results via three different kernels are shown in Figure 5. Normalized concatenated signals (when used as data input to SVM) gave the highest PCA feature reduction ratio. Figure 6 shows PCA feature reduction ratio for all three types of feature inputs to SVM

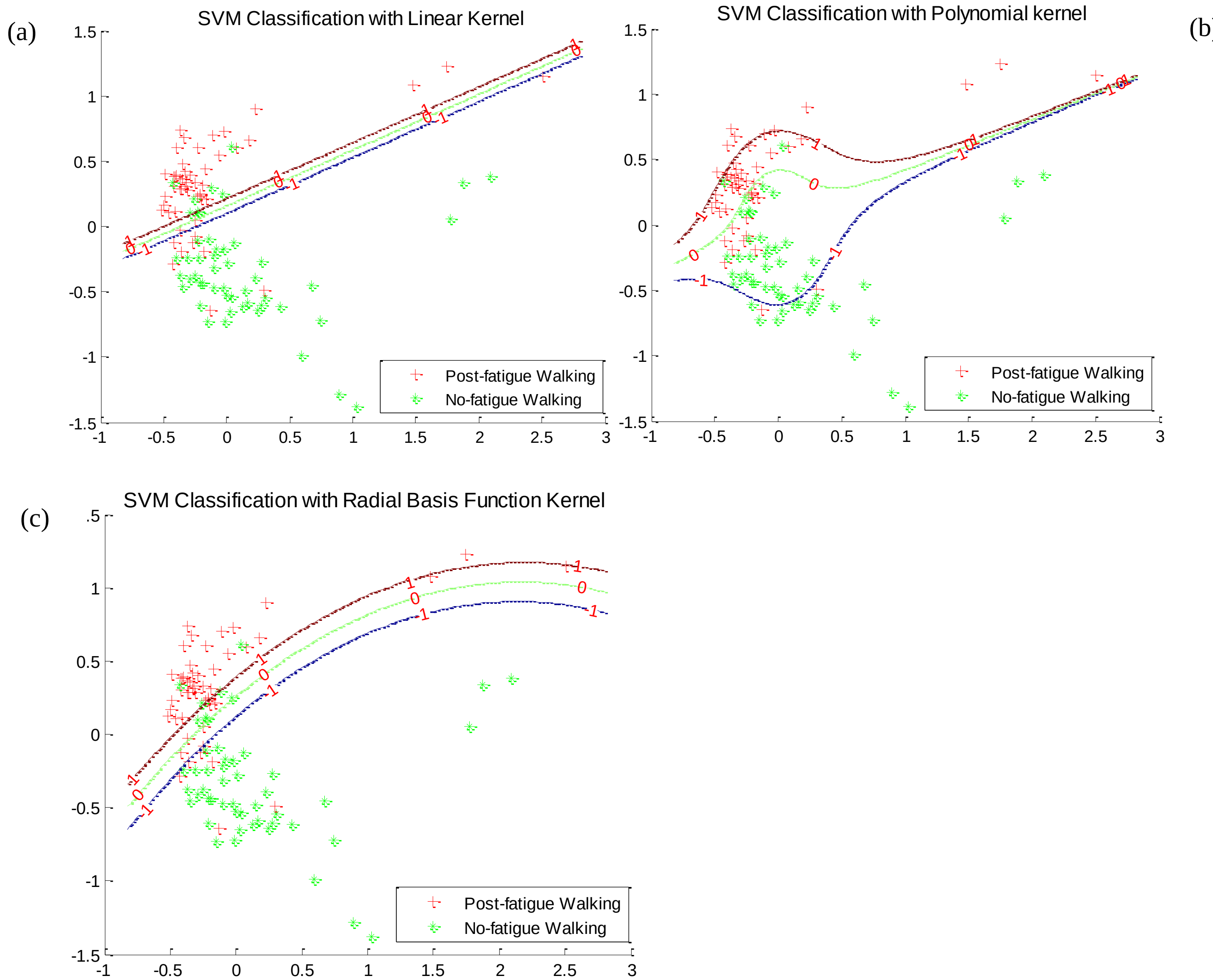

Figure 5: Inter-subject fatigue classification results via three different kernels: (a) linear kernel; (b) polynomial kernel, and (c) radial basis function kernel.

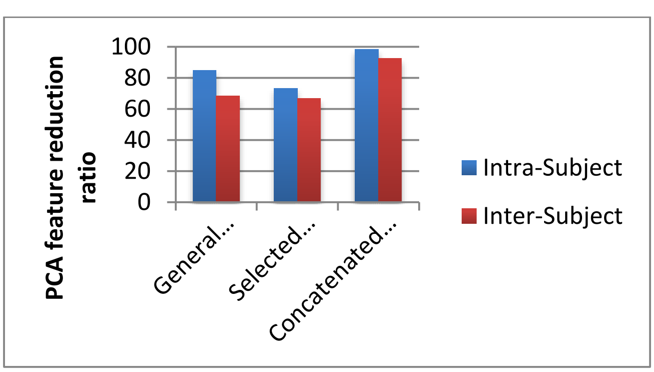

Figure 6: Intra and inter feature reduction ratio via PCA for three different feature set inputs ($PCA\ feature\ reduction\ ratio = \frac{Number\ of\ features - Principal\ Components}{Number\ of\ features} X\ 100$).

Discussion

In this study we explored the classification potential of SVM in recognition of gait patterns utilizing an inertial measurement unit (IMU) associated with lower extremity muscular fatigue. Our results indicate that fatigue effects are evident in individuals' gait patterns and extracted features and, SVM accurately classified fatigue/no-fatigue conditions. We found that SVM classifier incorporating trunk kinematic signals during gait has an excellent potential to predict fatigue status intra-individually (~97% accurate predictions) as well as inter-individually (~ 90% accurate predictions).

Our results indicate that fatigue adversely influences gait and this change, although subtle can provide helpful information for SVM to classify the status of lower extremity muscle fatigue. Our results suggest that single stance duration is decreased in post-fatigue walking trials. During stance phase of the gait cycle, proprioceptive input from extensor muscles and mechanoreceptors in the sole of the foot provide load information [11] to the central nervous system. Fatiguing of the muscles around a joint inhibits the joint's neuromuscular feedback and synergism between joint proprioception leading to instability and gait changes [15-23]. Thus, the reduced stance duration decreases foot-loading information through afferent sensory and proprioceptive mechanoreceptors, such as Golgi-tendon units, muscle spindles, and joint receptors, and may have adversely influenced motor control of the lower extremity during walking. Our results agree with Helbostad et. al. [9], who reported an increase in step width and trunk medio-lateral accelerations. Further, their study concluded that the variability in anterior-posterior acceleration is related to decreased walking stability [9]. The increased step width observed in post-fatigue walking may be associated with modulation of self-selected pace and loss of proprioception due to fatigue [56]. Further, our results affirm previously published literature documenting the variability in gait parameters in post-fatigue walking trials. Specifically, heel contact velocity (HCV) increased in post-fatigue walking trials [3]. Considering HCV is a kinematic gait parameter, which can drastically alter the friction demand (by change in required coefficient of friction) [51], an increase is acknowledged to increase the likelihood of slip-induced falls [3, 57, 58]. Thus our findings support previous literature by Helbostad et. al. [9] and Johnston et. al. [15], suggesting that lower extremity fatigue impairs gait performance and locomotor control.

Energy costs associated with walking may be utilized to classify fatigue conditions. Some researchers have modeled walking as an inverted pendulum system [59]. In this model, walker's body mass is assumed to be a point mass and, a rigid strut connects it to the point of ground contact. This point mass reaches the highest point at the middle of the stance phase [60-62]. The inverted pendulum model accurately predicts the general pattern of mechanical energy fluctuations of the body during walking [60-63]. Similarly in walking, accelerometer located at the trunk COM allowed the measurement of mechanical work done during walking (i.e., inducement of fatigue and its associated relationship to economy during walking as assessed by Jerk cost). Energy is defined as the external work done by muscles to maintain locomotion and is highly correlated with walking speed. In normal walking the kinetic energy and potential energy of the COM are nearly 180° out of phase whereas, in high speed walking, the fluctuations in gravitational potential energy are smaller than the fluctuations in kinetic energy [64]. In the inverted pendulum model, kinetic energy is transformed to potential energy from heel contact to mid-stance, in which the forward velocity of the COM decreases as the trunk arcs upwards over the stance foot. Subsequently, potential energy is transformed back to kinetic energy in the second part of the stance, from mid-stance to toe-off, with the COM moving downwards as the forward velocity of the COM increases. This energy exchange is similar to an oscillating pendulum and compliant with inverted pendulum model's emphasis on stiff-legged gait with the COM arcs on stance limb. An approach to minimize vertical movements of the COM was detailed by Inman and his colleagues [65], in which they identified several mechanisms involved in flattening the trajectory of the COM [65, 66], including sagittal plane knee flexion and extension during stance phase. Furthermore, Lee and Farley defined the change in distance between the COM and the point of ground contact using the virtual stance-limb compression; a measure influenced by (1) flexion of limb joints in the sagittal plane, (2) 3-D movement of pelvis and trunk, and (3) movement of the COM within the body [63]. Apropos to external work, localized muscle fatigue exhibits changes in sagittal kinematics of particular joints involved in fatigue protocol. Kellis and Liassou found that ankle muscle fatigue decreased ankle dorsiflexion, while knee fatigue increased knee flexion at initial heel contact [67]. Additionally, they reported increased hip extension following knee fatigue and increased plantar-flexion following ankle muscle fatigue, showing virtual stance limb length in post-fatigue walking is greater than that of a no-fatigue walking condition. Thus, compression of virtual stance limb plays a critical role in reducing the vertical displacement of the COM trajectory. Incidentally, following the inducement of fatigue, it appears that virtual stance-limb compression increases the upward movement of the COM, thereby increasing the pendulum-like energy exchange and subsequently decreasing stored elastic energy. Decrements in elastic energy storage in muscles, tendons, and ligaments as the ankle and knee flex, are prominent during the first half of stance phase; the elastic energy is utilized later to increase the COM velocity in the second half of stance phase. It's likely that elastic energy storage with post-fatigue walking becomes increasingly important due to decreased virtual stance-limb compression fostering higher energy exchange of kinetic and potential energy of the COM.

Hreljac and Martin concluded in their study that minimum jerk movements should also minimize energy consumption during walking [47]. When lower extremity muscle fatigue was induced, high abruptness in the COM acceleration (Jerk) is noticed and can be partially explained by decrements in flexion-extension angles [68]. Fatiguing of the muscles in lower extremity joints inhibits the joint's neuromuscular feedback and synergism between joint proprioception and muscular function leading to reduced stability [15-23] and stiffness [3, 69]. However, constant muscular stiffness has to be maintained to minimize jerk during repetitive, skilled movements [70]. It is seen that jerk at heel contact increased in magnitude by 2.3 folds and jerk cost increased by 2.8 folds in post-fatigue walking. Higher resultant accelerations (two folds higher range of acceleration) as well as higher signal energy magnitudes (5 folds higher signal energy) in post-fatigue walking trials were observed in our study. In essence, it appears that in post-fatigue walking trials, the total energy at the COM goes through large fluctuations during stance and the elastic energy storage is reduced due to lower virtual limb-stance compression; thus, resulting in higher energy dissipation through fluctuations in the trunk COM (similar to catching a baseball with extended elbow with greater impact (fatigued) vs. flexed elbow with less impact). It appears that SVM can map this non-linear inter-feature relationship using the kinematics of the trunk COM for better discrimination of fatigue and no-fatigue gait.
Previous researchers have adopted various gait feature extraction methods for SVM classification. Begg and coworkers differentiated elderly and young gait patterns using general features on minimum foot clearance data [29]. In another study, they selected kinetic and kinematic gait features for classification [31]. Whereas Eskoifer et. al. adopted concatenated waveforms from infrared markers to classify young and elderly gait [72]. Results of our investigation (Table 3 and Table 4) indicate that features extraction methods influenced classification accuracy. In inter-individual fatigue classification, general feature input performed with highest classification accuracy followed by selected feature input and concatenated waveform input. This performance in percentage of accuracies was also observed

for measuring tools i.e., forceplate and IMU. This may be attributed to higher inter-individual gait pattern differences [73], which are similar to findings by Eskoifer [72]. Additionally, in intra-subject fatigue classification, general feature input resulted in the highest classification accuracies, however, concatenated waveform input features had better classification accuracies than selected feature input. This may be attributed to intra-individual variability. In essence, general features exhibited superior classification accuracy and had important gait information to classify fatigue, on the contrary, the other two feature extraction methods lacked peculiar information relevant to achieving higher classification results.

Three different types of kernels were employed in SVM: linear, polynomial, and radial basis function. Both linear and RBF kernels performed well in intra-individual fatigue/no-fatigue classifications, which complied with Lee and Grimson's report [55], showing that linear kernel performs better than polynomial kernel in SVM gait recognition. Considering the computational cost, RBF and polynomial kernels need less time compared to linear kernels in the same conditions. As such, RBF kernel is the most promising kernel function in the fatigue classification schemes, and it may also provide better applicability to real time system implementation.

Limitations

We did not employ a separate cross-validation set for generalization of regularization parameter. One of the limitations of our study is that we did not employ any feature selection method to extract optimal set of features.

Conclusion

Inertial measurement units can assist in identification of localized muscle fatigue. Intra-subject fatigue classifications results in this study ranged from 93-98%, thus body worn sensors can potentially open doors for personalized monitoring on a regular basis in working environments to identify at-risk gait. SVMs are powerful machine learning tools applicable to the identification of post fatigue gait patterns by using a set of gait features related to the kinematics of the trunk COM during walking.
While the algorithms allow for online implementation, it is necessary to determine an optimal feature set that could automatically identify the most significant kinematic changes in gait after inducement of fatigue. Thus, we conclude that fatigue affects kinematics and gait characteristics, which can be assessed by an IMU using support vector machines.

Acknowledgements

This research was supported by the NSF (grant #CBET-0756058) and NSF-Information and Intelligent Systems (IIS) and Smart Health and Wellbeing -1065442 and 1065262. NIOSH (grant #CDC/NIOSHR01-OH009222). Additionally, supported by the NIH (L30-AG022963-04/NIH HHS/United States).

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4B. Aim 1: _Effects of Fatigue on Slip Initiation: Slip Propensity_

Parijat, P., and Lockhart, T.E., (2008), Effects of quadriceps fatigue on the biomechanics of gait and slip propensity. *Gait and Posture*, 28, 568-573.

Abstract

Existing epidemiological evidence suggests that localized muscle fatigue might be considered as an intrinsic risk factor that causes lack of balance control leading to falls. The goal of the present study was to examine how lower extremity fatigue (quadriceps) alters gait parameters which are indicative of slip propensity. Sixteen healthy young participants were recruited to walk across vinyl floor surface in two different sessions (Fatigue and No fatigue). Kinematic and kinetic data were collected using a three-dimensional video analysis system and force plates during both sessions. The results indicated a significant increase in the heel contact velocity, required coefficient of friction, and, a decrease in the transitional acceleration of the whole body center of mass, and peak average knee joint moment in the fatigue session. Additionally, joint kinematics showed increased knee flexion and reduced ankle dorsiflexion at the heel contact phase of the gait cycle during fatigue session. These findings provide new insights into the biomechanical relationship between localized muscle fatigue and gait parameters associated with slip propensity. The present study concluded that localized muscle fatigue affects gait parameters and hence can be considered as a potential risk factor for slip-induced falls.

Introduction

Slips and falls related injuries and fatalities continue to pose a significant burden to industry as well as the health sector. According to the Bureau of Labor Statistics (2003), 14% of accidental deaths in the workplace were reportedly caused by falls. In addition, slip recovery efforts have been shown to contribute to high rates of over-exertion injuries [1]. Floors, walkways or ground surfaces were the major extrinsic factors of fall accidents, causing over 86% of all fall-related injuries [2].

Occupationally induced localized muscle fatigue (LMF) has been recently identified as one of the major intrinsic factors contributing to slip and fall accidents [3, 4, 5]. The literature indicates that a third of the U.S. workforce must exert significant strength as part of the demands of their employment, and experience consequent fatigue at their work place [6]. Existing evidence provides convincing arguments that LMF of the lower extremity can disrupt the quality of the signal from the sensory inputs and increase the risk of slip-induced falls due to delayed response selection [5, 7, 30]. An increase in the postural sway was observed after fatigue from increased workload, which is associated with greater risks of falling [5, 16].

Previous studies have identified that, changes in the gait characteristics influence the risk of slip-induced falls [8, 9]. Increase in friction demand characteristics and heel contact velocity, along with reduction in transitional acceleration of the whole body center of mass during a gait cycle have been noted as risk factors for slip-induced fall accidents [8, 10, 25]. The lower limbs are essentially weight bearing structures that function to maintain stability and balance, and propel the body forward for locomotion, and hence LMF of lower extremities may have detrimental effects on the above-mentioned gait characteristics. More specifically, the importance of the knee joint musculature has been reported in terms of producing large flexion and extension moments while recovering from a slip [11]. Fatigue of the knee extensors and hip flexor muscles in a study demonstrated significant decreases in stabilization time compared with the fatiguing of other muscle group [12]. In addition, LMF of the quadriceps was found to adversely affect knee proprioception [13]. Proprioceptive information plays an important role in modifying the motor program to maintain dynamic stability during a slip perturbation [8], which might be compromised with LMF. The quadriceps and the hamstring musculature aid in control of knee flexion and extension and fatiguing these muscle groups may further influence the knee joint moment production and gait characteristics. While extrinsic factors leading to slip-induced fall accidents have been well documented and epidemiological studies link the incidence of slip-induced falls with muscle fatigue, the gait changes associated with muscle fatigue, and its relationship to slip-induced fall accidents still remain unclear.

The primary objective of the current study was to investigate the effects of LMF of the quadriceps muscle group on the gait characteristics. In addition, the relationship between gait parameter changes and higher risk of slip initiated falls was investigated. It was hypothesized that LMF of the quadriceps will adversely influence kinematic and kinetic parameters related to slip propensity leading to higher risks of slip-induced falls.

Methods

Subjects

Sixteen healthy young adults (10 males and 6 females) participated in the study. Informed consent was approved by the Institutional Review Board (IRB) of Virginia Polytechnic Institute and State University (IRB #03-069-AD) and was signed by all of the participants prior to the study. The participants (mean age 24.66 ± 3.58 years, Height 1.75 ± 0.07 m, Mass 65.86 ± 10.93 kg, BMI 22.14 ± 2.54 kg/m^2) did not have any musculoskeletal injuries that may affect their ability to perform the fatiguing exertions.

Equipment

Walking trials were conducted on a linear walkway (1.5 x 15.5m) embedded with two force plates (BERTEC # K80102, Type 45550-08, Bertec Corporation, OH 43212, USA). Kinetic data from the forces plate were collected at a sampling rate of 1200 Hz. Uniform experimental shoes were provided to participants to minimize shoe sole differences (Fig. 1a). Twenty-three reflective markers were placed over the various bony landmarks of the body [8]. A six-camera ProReflex system (Qualysis) was used to collect three-dimensional position data of the participant while walking. Kinematic data from the camera were sampled and recorded at 120 Hz. A fall arresting rig was used for safety [11]. A Biodex Dynamometer (Biodex Medical Systems, Inc., Shirley, NY) was used to induce fatigue. A special bilateral knee attachment was constructed for the Biodex that essentially worked the same as with one knee attachment. The attachment allowed the participants to extend and flex both of their knees together.

<u>Fatigue inducement and experiment protocol:</u> Bilateral fatigue was induced by performing repetitive isokinetic knee extension using the quadriceps. The fatigue inducement procedures were similar to those described by Yaggie and McGregor [14], with the exception that bilateral fatigue of the quadriceps was employed. All exertions were performed at 60º/s (eccentric-concentric mode), a value consistent with earlier fatigue protocols [15]. Participants were allowed to perform a 5-minute warm up on the Biodex before their MVE (maximum voluntary exertion) baseline measure was recorded. The MVE consisted of maximum voluntary extension of both the knees while applying minimal resistance when the joints returned to the original position. Participants performed the exertions repeatedly against a resistance set at 70 % of their baseline MVE. Visual feedback was provided to the participants for their current and

target moment levels. An MVE was performed at regular intervals (5 minutes) until the participants reached 60 % of their baseline MVE and this was considered as the fatigue state.

The participants were tested for two different sessions, fatigue (F) and no fatigue (NF) within a period of a week. The experiment was completely randomized for the two different conditions. During the first session (F or NF), participants were familiarized with the walking track and the Biodex. In the NF session, participants were instructed to walk at self selected pace across the linear walkway for 15-20 minutes. The force plate and kinematic data were then collected once the participants felt comfortable walking with the harness and produced consistent repetitive gait, i.e., the participant's feet landed correctly on the force plate and in the given sequence (right-left). Three gait cycles were recorded for each participant to represent the mean. In the F session, the participants first performed the walking trial similar to NF session and this was followed by the fatigue inducement procedure. The rationale for collecting the normal walking trials in the F session were to compare for differences in the normal gait between two separate days. Immediately after the fatiguing protocol (< 5 minutes), the participants were asked to walk across the walkway and data were collected to represent their fatigue walking. In order to avoid influence of fatigue recovery, the first trial was used for data analysis. Based on the existing evidence, full recovery of the stabilometric measures is expected within about 20 minutes but this holds true for postural sway studies after fatigue [14, 16]. To avoid confounding effects due to recovery from imposed muscle fatigue while walking, it is important to minimize the time between fatiguing exercise and data collection. To assess the recovery time after fatigue, a pilot test was conducted prior to the experiment. In the pilot test, six healthy young adults (3 males and 3 females) performed the fatigue exertions as described above and then they were asked to walk on the linear walkway once. Immediately after walking, they were brought back to the Biodex and their MVE was measured again. After the MVE measurement, they were asked to walk again. This was repeated until the participants recovered their baseline MVE. The study indicated different recovery patterns for each individual with no one returning to their baseline MVE before 10 minutes (Fig. 1). Therefore, results from the pilot study were utilized as a reference time window for data collection after fatigue inducement to reduce the effects of fatigue recovery.

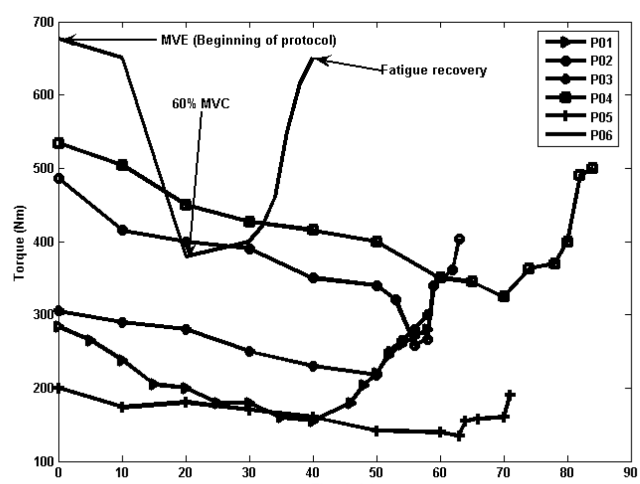

Figure 1. Bilateral isokinetic fatigue development and recovery pattern of six young healthy adults. (demonstrates, baseline MVE- 0 minute, end of fatigue protocol at 60% MVE, and recovery period after fatigue inducement).

Data Analysis

The independent variable in this experiment was the fatigue status (F and NF). The dependent variables consisted of various kinematic and kinetic gait variables. Heel contact velocity (HCV) was calculated by numerically differentiating the position data of the heel before and after the heel contact phase of the gait cycle [8, 17]. Transitional acceleration of the whole body center of mass (TA) was defined as the change in the horizontal center of mass (COM) velocity between the heel contact phase and shortly after the heel contact phase of the gait cycle [8]. Walking velocity (WV) was obtained from the whole body COM velocity during forward progression [8]. The joint angles and angular velocities were determined using the position data as described by Liu and Lockhart [11]. The required coefficient of friction (RCOF) is defined as the ratio of horizontal ground reaction force to vertical ground reaction force. It represents the minimum required coefficient of friction between the shoe and floor interface to prevent slipping [18]. Two-dimensional sagittal knee joint moment was calculated using the inverse dynamics approach [11]. The joint moment was normalized to body weight for data analysis.

Table 1. Summary of kinematic and kinetic gait parameters (HCV-heel contact velocity, WV- walking velocity, RCOF-required coefficient of friction, TA- whole body COM transitional acceleration, Kneemom_{peak}- knee moment peak P1 and P3) during the no fatigue (NF) and fatigue (F) walking trials * $p < 0.05$, ** $p < 0.01$

Variables	Session		ANOVA
	NF	**F**	
	mean(S.D)	*mean(S.D)*	
HCV (cm/s)	81.94 (51.22)	97.83(66.67)	*
WV (cm/s)	127.02(14.42)	119.29(20.32)	N.S
RCOF	0.15(0.007)	0.18(0.005)	*
TA(cm/s^2)	199.21(41.27)	159.27(57.16)	*
Knee angle at heel contact (deg)	8.47 (1.34)	2.59(1.50)	**
Ankle angle at heel contact (deg)	93.77 (4.74)	86.66(3.95)	**
Knee angular velocity at heel contact (deg/s)	235.09 (15.31)	134.07 (8.58)	*
Ankle angular velocity at heel contact (deg/s)	56.18 (2.41)	69.60 (2.51)	*
Kneemom_{peak} (Nm/kg) P1	0.47(0.19)	0.37(0.22)	N.S
Kneemom_{peak} (Nm/kg) P3	0.65(0.36)	0.43(0.15)	**

The peak value of the knee joint moment was utilized for the analysis. All of the dependent variables were analyzed for the stance phase of the gait cycle. A within-subject design was utilized wherein the same subjects repeated the two treatment conditions (F and NF). A one-way repeated measures ANOVA was used to test for significant differences between various gait parameters during NF and F sessions. The level of significance was set at $\alpha < 0.05$. Bivariate analysis was performed to examine the correlation between the different dependent measures.

Results

The one way ANOVA indicated that the participants walked with a higher HCV in the F session ($F_{(1,31)} = 33.86$, $p = 0.01$). The TA in the forward direction was reduced after the fatiguing protocol ($F_{(1,31)} = 3.85$, $p = 0.04$). Participants walked with a reduced speed after fatigue exertions but the differences were not statistically significant ($F_{(1,31)} = 1.52$, $p = 0.08$). The mean and standard deviation with the statistical results is provided in Table 1.

At heel contact, the ankle and knee kinematics were significantly different for NF and F sessions (ankle-$F_{(1,31)} = 18.01$, $p = 0.009$, knee-$F_{(1,31)} = 130.02$, $p = 0.0001$) (Table 1). The knee was flexed to a greater extent at the heel contact phase of the gait cycle during F session as compared to the NF session (Fig. 2a). The gait profile however was similar in both of the sessions, with the knee flexed for around 30 % of the stance and then flexed again during terminal stance. The ankle was in dorsiflexion at the heel contact but rapidly reached its peak plantar flexion around 15% of the stance phase for NF session (Fig. 2a). During F session, the ankle followed a similar gait profile but with reduced dorsiflexion and plantarflexion. The knee and ankle angular velocities were significantly different at the heel contact phase of the gait cycle (Fig. 2b, Table 1).

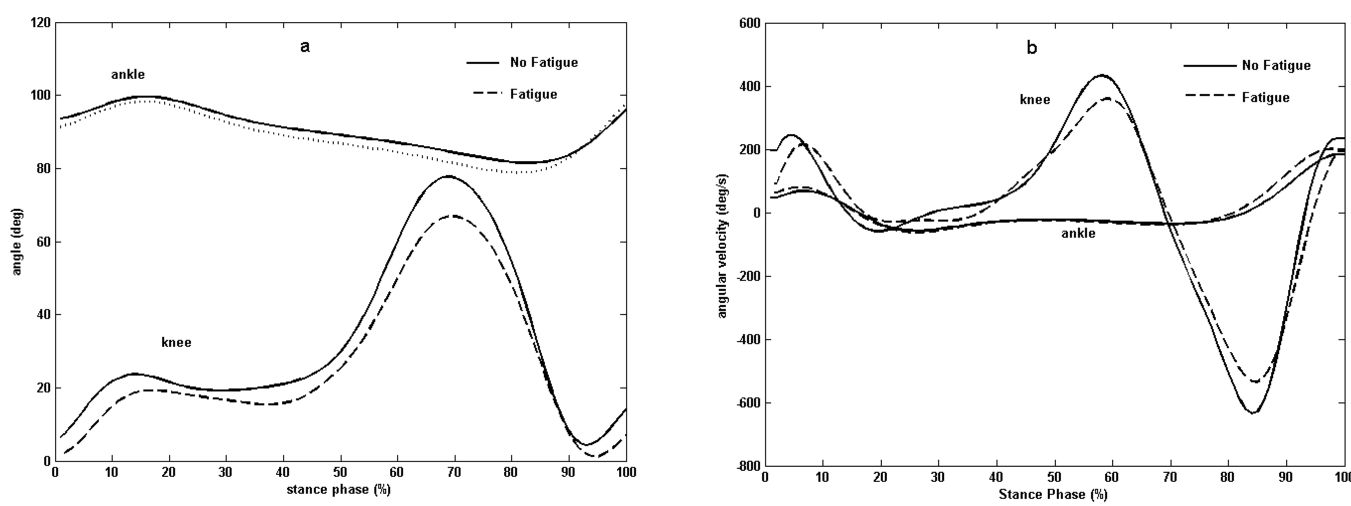

Figure 2(a). Mean profile of ankle and knee angles during the no fatigue and fatigue walking trials (b). Mean profile of ankle and knee angular velocities during the no fatigue and fatigue walking trials (+ dorsiflexion (ankle), + extension (knee)), (0- heel contact, 100- toe off).

Consistent patterns of RCOF were observed during both of the sessions in all of the participants (Fig. 3). It was also observed that RCOF was higher during F session ($F_{(1,31)} = 9.73$ $p = 0.04$) as compared to the NF session. The sagittal knee joint moment showed flexor and extensor moment alternatively in both NF and F sessions (Fig. 4). Two distinctive extensor moment peaks (P1 and P3) were analyzed for significant differences between NF and F sessions. Although not statistically significant ($p>0.05$), there was a decrease in the peak extensor knee joint moment P1 in the F session as compared to the NF session (Table1). The Peak 3, which is also extensor dominant, was decreased in the F session and was significantly different from NF session ($F_{(1,23)} = 16.89$, $p = 0.002$) . A second peak (P2) was observed in case of F session which was flexor dominant and occurred during 60-70% of the stance phase. However, in the NF session the joint moment profile was different with the knee less extended but not fully flexed at 60-70% of the stance phase. The bivariate correlation between the different dependent variables indicated that RCOF was positively correlated to HCV($r = 0.39$, $p = 0.026$) and TA was positively correlated to the peak knee joint moment (P1 ($r = 0.31$, $p=0.03$), P3 ($r = 0.45$, $p= 0.02$)). Additionally, there was a negative correlation between HCV and TA ($r = -0.37$, $p = 0.03$). Although the correlations between the variables were significant, they were overall weak.

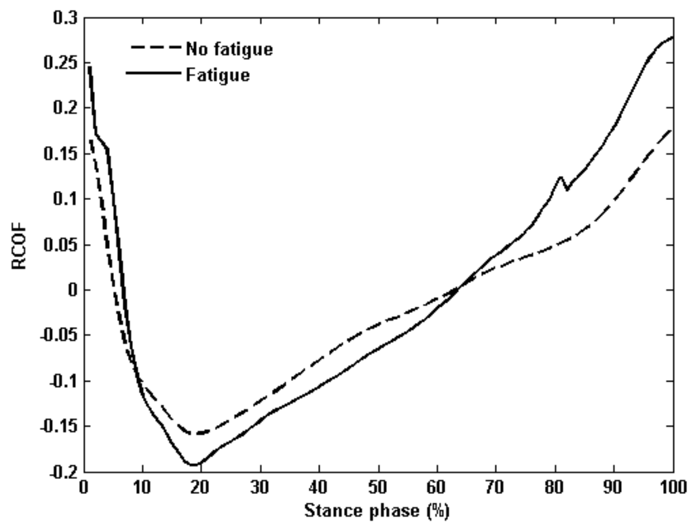

Figure 3. Ensemble average of the required coefficient of friction (RCOF) during the no fatigue and fatigue walking trials. (0- heel contact, 100- toe off).

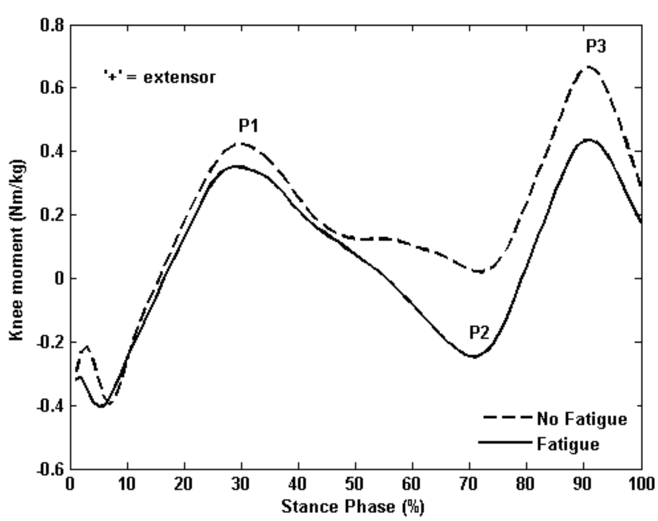

Figure 4. Average knee joint moment profile at the stance phase of gait cycle during the no fatigue and fatigue walking trials. (0- heel contact, 100- toe off).

Discussion

The purpose of the study was to investigate if LMF will influence gait characteristics and increase the risk of slip-induced falls. The results indicated that participants walked with a higher HCV after fatiguing the quadriceps. HCV is considered an important gait parameter related to friction demand and slip-induced falls. It has been studied that HCV affects the friction demand characteristics by altering the ratio of horizontal to vertical foot forces [8]. Increases in HCV during a critical time of weight transfer may increase the potential for slip-induced falls if the friction between the heel and the floor is reduced due to contamination of floor surface. Additionally, investigations of older adult's gait have revealed that the risk of slip-induced falls were higher due to an increase in HCV [19, 20]. The results are consistent with the study by Saggini et al. [21] who examined the effects of lower extremity fatigue and concluded that fatigue increased the gait cycle time and also increased the horizontal heel velocity. The bivariate analysis suggested a positive correlation between HCV and RCOF, indicating that the friction demand characteristics increased with an increase in the HCV. A likely factor influencing the higher HCV may be a change in the quadriceps-hamstrings co-activation rate [8]. Co-activation of hamstring and quadriceps muscles is important in heel contact dynamics. Fatigue of the quadriceps may influence this process and thereby increase the HCV. Although implicated, further EMG study is required to understand the relationship between quadriceps fatigue and the co-activation dynamics. In this study, we did not control the walking velocity of the participants. Reduction in WV has been identified as a risk factor for slip-induced falls in elderly [10]. Although implicated, no significant difference in WV was observed in this study. This might be attributed to the differences in individual strategy to adapt their gait speed after fatigue.

The forward momentum of the whole body COM is important during slip initiation and recovery phases. TA is an important parameter in assessing this forward momentum of the body. It was observed in this study that after fatiguing the quadriceps musculature, the TA was reduced. This is in agreement with past studies where it was observed that fatigue affected the back acceleration and changed the loading rate during the gait cycle [22]. Lockhart et al. [8] indicated that reduced push-off force of the stance leg further reduced TA and increased RCOF and risk of slip initiation. This indicates that a reduction in the TA due to LMF is likely to increase the friction demand at the shoe floor interface of the contacting foot, thus increasing the risk of slip initiated falls.

The joint kinematics were significantly different for F and NF sessions during the heel contact phase of the gait cycle. After the fatiguing protocol, participants had a greater knee flexion at the heel contact and less knee extension during terminal stance. This can be expected, as fatigue of the quadriceps will affect the knee extensors. In addition, fatigue of the quadriceps affected the ankle kinematics with less dorsiflexion at the heel contact in the F session. The knee and ankle angles during the NF session are in accordance with the previous literature [17, 23].

In terms of kinetic gait parameters, the results indicated a higher friction demand during F session. The friction demand characteristics have been implicated as an important predictor variable related to severity of falls [8, 24]. It has been observed that the onset of lower extremity fatigue during walking changed the loading rate and increased the ground reaction forces [25]. As RCOF is dependent on the ground reaction forces (horizontal and vertical), this would mean that increased ground reaction forces due to fatigue will alter friction demand characteristics. Furthermore, Lockhart et al. [8] indicated that a reduction in TA is likely to increase the friction demand at the shoe floor interface of the contacting foot. Increased initial friction demand (i.e., RCOF) would lead to a higher likelihood of slips associated with low coefficient of friction floor surfaces. It has been observed previously that both TA and HCV were predictor variables for RCOF in younger adults [10]. This is in accordance with the current study.

The average knee joint moment profiles during the NF session are in accordance with previous literature [11, 23]. The knee joint moment profile during the F session had reduced extensor and increased flexor peak moment. The peak extensor joint moment (P1) was less after fatigue, though the difference was not statistically significant for 30% of the stance phase. After midstance, there was a flexion moment peak (P2) in the F session and the knee extension moment was significantly reduced during the push off phase of the gait cycle. It has been indicated previously that reduced push-off force of the stance leg further reduced the transitional acceleration of the whole body COM and increased RCOF and risk of slip initiation [8]. The knee joint moment in the F session are contradictory to some studies in the past [26, 27]. The discrepancies may be attributed to variation in fatigue patterns (i.e., general lower extremity fatigue, dorsiflexor fatigue) in the previous studies.

One of the limitations of the present study was that each participant reached their fatigue level at different times. These limitations can affect the results due to the difference in the fatigue level in each individual. Although implicated, the assumption of 60% of baseline MVE as fatigue state prior to testing ensured that all participants were fatigued at similar levels. Previous studies on the difference between the strength in unilateral and bilateral exertions have produced equivocal results [28]. However, a study by Jacobi et al. [29] concluded that the force production is not altered during bilateral contractions. Bilateral fatigue was induced in the current study to simulate a more realistic scenario where both limbs are fatigued (i.e., walking).

Future research will investigate the effects of LMF of other musculatures in the lower extremities (i.e., hamstrings, ankle plantar flexors) on normal gait and slip events in a real world situation. However, results from the present study can be used as preliminary information on the specific gait parameters that are affected by LMF. The present study provides evidence that there is a relationship between LMF of the quadriceps and the gait parameters, which are linked with higher risks of slip-induced falls.

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4C. Aim 1: ***Effects of Fatigue on Slip Initiation (Ankle Fatigue)***

Introduction

Slips and falls related injuries and fatalities continue to negatively impact occupational environments. Along with slip-induced fall outcomes, slip recovery efforts have also been shown to contribute to high rates of over-exertion injuries (Courtney & Webster, 2001). Floors, walkways or ground surfaces are a major extrinsic factor for fall accidents, causing over 86% of all fall-related injuries (Yoon & Lockhart, 2006). Occupationally induced localized muscle fatigue (LMF) has been identified as one of the intrinsic factors contributing to slip and fall accidents (Bentley & Haslam, 2001; Cohen & Lin, 1991; Hsiao & Simeonov, 2001). One third of the US workforce exerts considerable strength as part of the demands of their employment and experience consequent fatigue at work (Swaen et al., 2003).

Physical exertion in lower extremity may cause temporary localized muscle fatigue 1in ankle dorsiflexors and plantarflexors and thus reduce strength in these muscle groups and influence gait and balance. LMF of these muscles can also disrupt the quality of the signal from the sensory inputs and increase the risk of slip-induced falls due to delayed response selection (Hsiao & Simeonov, 2001; Lipscomb et al., 2006; Sparto et al., 1997). Generally, an increase in postural sway has been observed after fatigue from increased workload, which appears to be associated with a greater risk of falling (Hsiao & Simeonov, 2001; Nardone et al., 1997).

More specifically, ankle dorsiflexors and plantarflexors muscles play an important role in maintaining balance during standing (Horak et al., 1989; Laughton et al., 2003). Multiple studies have reported deterioration in postural balance during quiet standing following plantar-flexor muscle fatigue (Ledin et al., 2004; Noda & Demura, 2006; Roerdink et al., 2011). It has been found that ankle plantarflexor fatigue increased sway area and velocity particularly in closed eyes condition(Boyas et al., 2013), and shifted the mean COP position in posterior direction probably an adaptation to increase postural stability(Boyas et al., 2013). Further, compensate for the effects of fatigue participants generally increased the ankle and hip flexion angles(Boyas et al., 2013). Indeed, ankle musculature plays an important role in maintaining balance during single and bilateral limb standing (Billot et al., 2010; Riemann et al., 2003), also corrections of postural balance in anterior postural plane(Schieppati & Nardone, 1999). The strong reliance on ankle strategy is maintained and even gets strengthened with fatigue (Boyas et al., 2013), which further emphasizes on important role of ankle muscles even in fatigue state.

Changes in the gait characteristics such as increase in friction demand characteristics, heel contact velocity and reduction in transitional acceleration of the whole body center of mass during a gait cycle have been noted as risk factors for slip-induced fall accidents (Lockhart et al., 2003). Previous studies on effects of fatigue have suggested that LMF does not influence ground reaction force (Reinschmidt et al., 1994; Willson & Kernozek, 1999), but affect plantar pressure distribution while walking in fatigued state than in normal walking condition. Moreover, plantar-flexor muscle fatigue is associated with decreased ankle proprioception and degradation of force sense at the ankle joint (Vuillerme & Boisgontier, 2008), and also degraded the sense of limb position (Vuillerme et al., 2007). Although epidemiological studies link risk of slip-induced falls with LMF (Hsiao & Simeonov, 2001; Lundin et al., 1993), the changes in biomechanical gait parameters associated with fatigue, and its relationship to slip-induced fall accidents still remain unclear.

The primary objective of the current study was to investigate the effects of localized muscle fatigue of the plantarflexor muscle group on gait characteristics. The relationship between changes in gait variables and higher risk of slip initiated falls was investigated. It was hypothesized that localized muscle fatigue of the ankle plantarflexors would affect the kinematic and kinetic variables related to slip propensity leading to higher risks of slip-induced falls.

Materials and Methods

Participants

15 healthy young adults participated in this study. Respective means (SD) of participant age, stature, and body mass were 36.9 (11.6) yrs, 176.9 (7.4) cm, and 73.7 (14.3) kg. All participants were healthy, independent and non-sedentary and, were formally screened for major musculoskeletal, cardiovascular, and neurological disorders by a research coordinator during initial participant contact. Exclusion criteria of this study were factors that could interfere with gait, such as medication use, presence of neuromuscular disease and, balance and vision disorders. Informed consent was approved by the Institutional Review Board (IRB) of Virginia Tech and was signed by all participants prior to the study.

Instrumentation and Experimental Procedure

Participants were instructed not to perform any strenuous exercise 48 hours prior to the experiment. All experiments were conducted between 11 AM and 4 PM to control the circadian effects of fatigue. Walking trials were performed both prior and after the fatigue inducement. Participants walked on a linear walkway (15.5 X 1.5 m) embedded with two force plates (BERTEC #K80102, Type 45550-08, Bertec Corporation, Columbus, OH 43212, USA) in the middle of the walkway. A six-camera ProReflex system (Qualysis) was used to collect three-dimensional movement data of participants using passive infra-red markers. A total of 32 reflective markers were placed over bony landmarks of the foot, shank, thigh, pelvis, trunk, and head. To ensure safety, during all trials participants wore a fall arrest harness that was connected to an overhead rail. To minimize variability due to shoe sole properties, a consistent type of shoe was used by participants during the experiment. The data was processed using custom software written in MATLAB (version 7.11.0, 2010, computer software, The Math- Works Inc., Natick, Massachusetts).

All non-fatigue walking trials were preceded by acclimatization in laboratory environment and warm-up for about 10 minutes (walking back and forth three times around the laboratory). Practice trials were performed with the dynamometer followed by data collection for maximum voluntary torque and normal walking. Before inducing fatigue warm up exercises were conducted. Participants were considered fatigued when their isokinetic MVC was below 70% of initial isokinetic values for all three trials.

Non-fatigue walking trials

Participants were instructed to walk at their self-preferred pace on the walking track and gait characteristics were assessed in the middle portion (5 m) of this walkway. Infrared markers on both feet were used to determine step length, step width, heel contact velocity and single stance time. Step length refers to the linear distance in the direction of progression between successive points of foot-to-floor contact of the first foot and contralateral foot. The step length is calculated from the difference between consecutive positions of the heel contacting the floor. The step width is the distance between the rear end of the right and left heel centerlines along the mediolateral axis of foot. Heel contact velocity is the instantaneous horizontal heel velocity at heel contact is calculated utilizing heel velocities in the horizontal direction at the foot displacement of 1/60 s before and after the heel contact phase of the gait cycle. Single stance time refers to the time person is standing on one foot.

Reflective marker on the sacrum was used to determine walking velocity. Heel contact and toe-off time events were confirmed using ground reaction forces measured (sampling frequency 1200Hz) from the force plates positioned across the center area of the walkway. Ground reaction force measurement was reviewed in every trial for ascertaining foot placement in the desired sequence (i.e., left-right heel contacts on the two forceplates). If the foot placement did not lie within the force platform boundary, the participants were requested to repeat the trial. Five good walking trials were collected for each experimental condition and each trial consisted of 6-7 complete gait cycles.

Fatigue Inducement

A custom built Biodex (Biodex System 3 Dynamometer, Shirley, New York, USA) attachment for the ankle was used to assess maximum voluntary isokinetic exertions (MVE) during ankle dorsiflexion and plantarflexion. The Biodex attachment was designed to measure ankles torques of both feet. The participant seated on the biodex chair in a recumbent position with the knee in 90 degrees of flexion, and the foot secured to the footplate attached to the dynamometer. The participant's thigh and pelvis were immobilized using the Velcro straps.

The total range of motion of ankle joint was not employed in dynamometer to minimize the injury risk as well as to have range of motion comparable with that used during walking. Laboratory footwear were used by the participants and an angle of 90° of ankle joint was defined as the angle when sole of the shoe was perpendicular to the axis of the lower leg and 80°was in the direction of dorsiflexion. Foot was strapped on the foot device (Biodex Systems), which was a metal pedal. The pedal was carefully adjusted under the metatarsophalangeal.

Firstly, warm-up exercises were performed which consisted of 4 sets of 2 isokinetic plantar flexion contractions at 60° per second. Participants produced intensity levels of 25%, 50%, 75%, and 100% of perceived MVE for each of the 4 sets (Hortobagyi et al., 2001; King et al., 2012). All MVEs were performed using the ankle attachment with a dynamometer. Isokinetic MVE was defined as the maximum plantarflexion torque produced during 3 trials of 10 second isokinetic ankle plantar flexion task performed at 60° per second. The ankle fatigue inducement exercises were performed using the same attachment in isotonic mode set to 70% of initial MVE for plantar flexion and 15-30% for dorsi-flexion. The different plantar-flexion and dorsi-flexion resistance exercise torques were adopted for repeated eccentric-concentric muscle actions because dorsiflexors exhausted earlier than plantarflexors (Svantesson et al., 1998) (potentially due to irradiation effect (Todor & Lazarus, 1986)). The participants repeatedly performed isotonic contractions at 22 repetitions per minute. An exercise was set for 5 minutes and was followed by measurement of three isokinetic MVEs using dynamometer. Experimenters did not instruct participants to take break between the exercise sets, but it was kept on participant's choice to start their next new exercise set as soon as they felt ready. The exercise sets continued until the participants reached 70% of their baseline MVE for all 3 consecutive contractions; this was categorized as fatigued state (time taken by participants was 45±7 minutes to reach this state).

Fatigue walking trials

After inducement of fatigue as determined by degradation in MVEs, locations of all infrared markers were re-checked. Participants did not warm-up after isotonic fatiguing exercises but were asked to walk again on the walking track at their own preferred pace and fast walking pace. All gait characteristics were derived similar to that mentioned in non-fatigue walking section and six good walking trials which included three at preferred walking pace and three at fast walking pace were collected. The complete experiment lasted for 3-4 hours.

Recovery from fatigue and walking trials

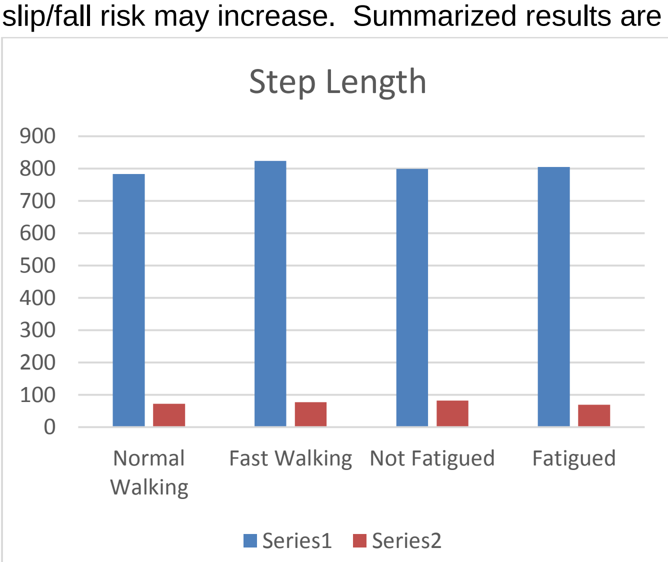
After collecting six walking trials (3 preferred normal pace and 3 fast walking pace) participants were strapped with ankle attachment and MVEs were collected. Participants then performed the walking trials on the track and the whole process repeated for about 45 minutes with mentioned strength measurements in between after each 7 minutes. These inter trial MVEs depicted recovery from fatigue with elapse of time between measurements.

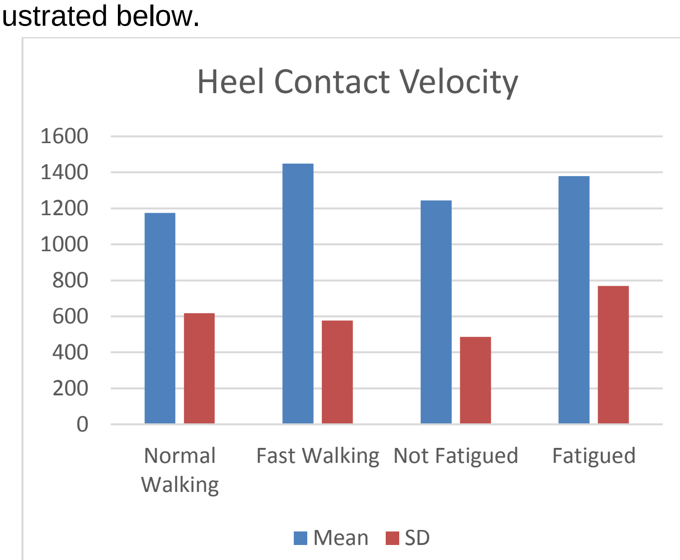
Statistical data analysis

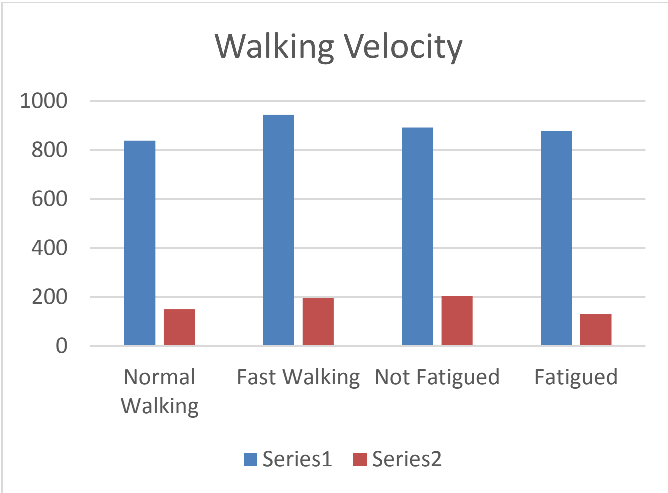
Gait parameters such as step length, step width, heel contact velocity, and single stance time were computed for all five trials around the two forceplates. The level of significance was set at $\alpha < 0.05$. To determine the relationship between various gait variables to walking speed and fatigue condition, we fit generalized estimating equation (GEE) models that account for the within subject correlation among each subject's trials. We selected the compound symmetry covariance structure as the most appropriate structure for our data after comparing several models using the Akaike information criterion. Model assumptions were validated using the distributions of the residuals for each model. A 0.05 significance level was used throughout this analysis. All analysis was performed in SAS 9.4 (SAS Institute Inc., Cary, NC, USA).

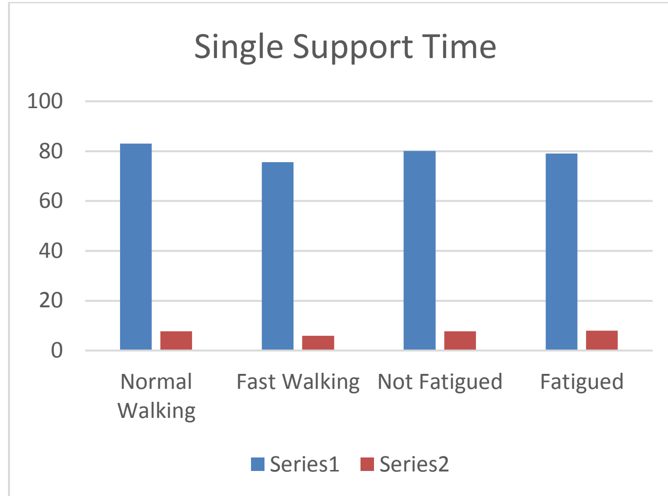
Results

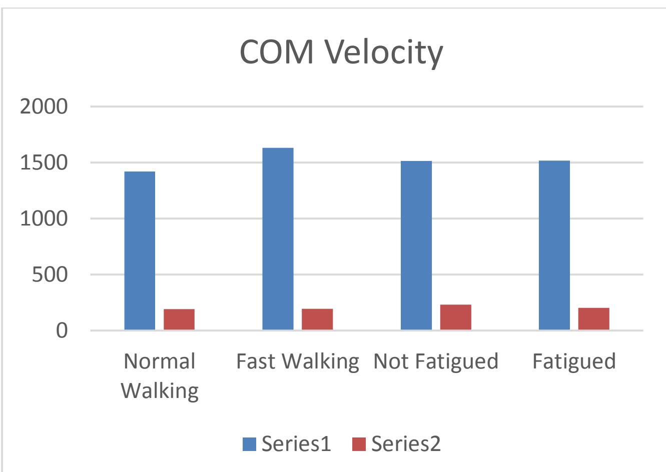
Overall, the fatigue effects were observed on various gait parameters. Specifically, heel contact velocity and walking velocity was faster when fatigue with longer step length. Additionally, single support time was reduced with decreased toe clearance height. It appears that when fatigued, we may revert to more safe gait – i.e., decreasing single support time to increase the double support time for better dynamic balance maintenance. In order to do so, individual may needed to walk faster with higher heel contact velocity. Although RCOF was not significantly affected by fatiguing, given a slippery floor surfaces, slip/fall risk may increase. Summarized results are illustrated below.

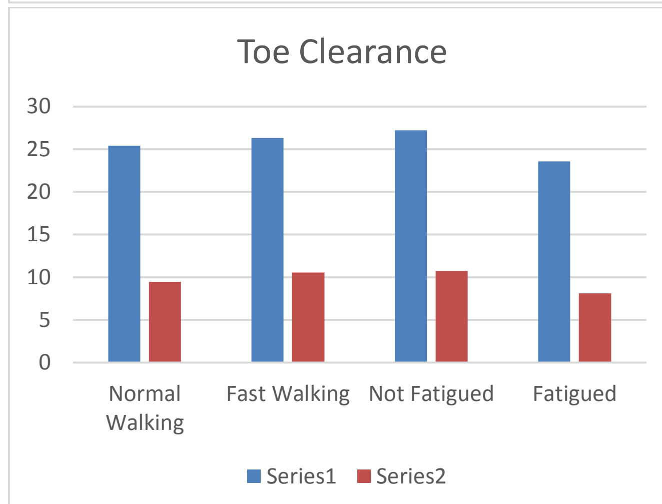












Analysis Of GEE Parameter Estimates: StepLength

Empirical Standard Error Estimates

Parameter			**Estimate**	**Standard Error**	**95% Confidence Limits**		**Z**	**Pr > |Z|**
Intercept			776.8678	18.1917	741.2127	812.5229	42.70	<.0001
FatigueState	Post 1		21.0045	10.7028	0.0275	41.9815	1.96	0.0497
WalkType	F		52.6195	14.3347	24.5241	80.7149	3.67	0.0002
FatigueStat*WalkType	Post 1	F	-37.1016	13.3357	-63.2391	-10.9641	-2.78	0.0054

Analysis Of GEE Parameter Estimates: StepLength								
Empirical Standard Error Estimates								
Parameter			**Estimate**	**Standard Error**	**95% Confidence Limits**		**Z**	**Pr > |Z|**
Intercept			776.8678	18.1917	741.2127	812.5229	42.70	<.0001
FatigueState	Post 1		21.0045	10.7028	0.0275	41.9815	1.96	0.0497
WalkType	F		52.6195	14.3347	24.5241	80.7149	3.67	0.0002
FatigueStat*WalkType	Post 1	F	-37.1016	13.3357	-63.2391	-10.9641	-2.78	0.0054

Analysis Of GEE Parameter Estimates: HCV								
Empirical Standard Error Estimates								
Parameter			**Estimate**	**Standard Error**	**95% Confidence Limits**		**Z**	**Pr > |Z|**
Intercept			1071.229	58.2044	957.1507	1185.308	18.40	<.0001
FatigueState	Post 1		267.3430	129.2811	13.9566	520.7293	2.07	0.0386
WalkType	F		382.6577	100.0631	186.5377	578.7777	3.82	0.0001
FatigueStat*WalkType	Post 1	F	-297.896	180.4411	-651.554	55.7625	-1.65	0.0988

Analysis Of GEE Parameter Estimates: Walkingvelocity								
Empirical Standard Error Estimates								
Parameter			Estimate	Standard Error	95% Confidence Limits		Z	Pr > |Z|
Intercept			829.8021	34.3567	762.4643	897.1399	24.15	<.0001
FatigueState	Post 1		34.4092	16.2768	2.5073	66.3111	2.11	0.0345
WalkType	F		141.9716	28.8628	85.4016	198.5417	4.92	<.0001
FatigueStat*WalkType	Post 1	F	-108.812	38.7663	-184.793	-32.8317	-2.81	0.0050

Analysis Of GEE Parameter Estimates: V_COM								
Empirical Standard Error Estimates								
Parameter			Estimate	Standard Error	95% Confidence Limits		Z	Pr > |Z|
Intercept			1414.431	47.7656	1320.812	1508.050	29.61	<.0001
FatigueState	Post 1		45.8425	21.1863	4.3182	87.3668	2.16	0.0305
WalkType	F		238.0110	40.7316	158.1786	317.8434	5.84	<.0001
FatigueStat*WalkType	Post 1	F	-91.3625	27.0880	-144.454	-38.2709	-3.37	0.0007

Analysis Of GEE Parameter Estimates SST1								
Empirical Standard Error Estimates								
Parameter			**Estimate**	**Standard Error**	**95% Confidence Limits**		**Z**	**Pr > |Z|**
Intercept			83.7623	1.9748	79.8918	87.6328	42.42	<.0001
FatigueState	Post 1		-2.2382	0.9849	-4.1686	-0.3078	-2.27	0.0231
WalkType	F		-8.1785	1.3249	-10.7753	-5.5817	-6.17	<.0001
FatigueStat*WalkType	Post 1	F	1.6571	1.3993	-1.0855	4.3998	1.18	0.2363

Analysis Of GEE Parameter Estimates: Toe Clearance								
Empirical Standard Error Estimates								
Parameter			**Estimate**	**Standard Error**	**95% Confidence Limits**		**Z**	**Pr > |Z|**
Intercept			27.0726	2.5132	22.1469	31.9984	10.77	<.0001
FatigueState	Post 1		-3.8562	1.5352	-6.8650	-0.8473	-2.51	0.0120
WalkType	F		1.1426	1.0262	-0.8688	3.1539	1.11	0.2655
FatigueStat*WalkType	Post 1	F	0.5011	1.5953	-2.6256	3.6279	0.31	0.7534

Analysis Of GEE Parameter Estimates: RCOF								
Empirical Standard Error Estimates								
Parameter			**Estimate**	**Standard Error**	**95% Confidence Limits**		**Z**	**Pr > |Z|**
Intercept			0.1801	0.0063	0.1677	0.1925	28.47	<.0001
FatigueState	Post 1		0.0018	0.0037	-0.0055	0.0091	0.47	0.6351
WalkType	F		0.0115	0.0034	0.0048	0.0183	3.37	0.0008
FatigueStat*WalkType	Post 1	F	-0.0072	0.0078	-0.0225	0.0081	-0.92	0.3550

Disscussion

Localized muscle fatigue of the lower extremities induced by prolonged walking influences initial gait characteristics such as step length, heel contact velocity, and ground reaction forces. Saggini et al., (1998) examined the effects of localized muscle fatigue on the lower extremity and concluded that fatigue increased the gait cycle time and also increased the heel contact velocity. Initial gait characteristics such as heel contact velocity can alter the ratio of horizontal to vertical foot forces (Lockhart et al., 2003). Increases in horizontal heel velocity during a critical time of weight transfer may increase the potential for slip-induced falls if the friction between the heel and the floor is reduced due to contamination of the floor surface. For example, investigations of older individuals' gait characteristics by Lockhart et al. (2000) and Winter (1990) revealed that the risk of slip-induced falls was higher due to the higher heel contact velocity. A likely factor influencing the higher horizontal heel contact velocity may be a decrease in hamstring activation rate due to localized muscle fatigue. These muscles become active at the termination of swing phase, being elongated as they act to decelerate the swing leg, and help extend (control) the knee and hip. As discussed earlier localized muscle fatigue is defined as an acute impairment in the ability to exert force or power, and fatigue occurs as the metabolic by-products reduce intra-muscular conduction velocity until the muscles become unable to produce the desired forces (Svantesson et al., 1998; Gefen et al., 2002). Fatigue-induced contractile process and excitation-contraction coupling failure may decrease the hamstring activation leading to higher heel contact velocity. Thus, effects of localized muscle fatigue on gait characteristics and specifically higher horizontal heel contact velocity during the critical phase of the gait cycle may increase the likelihood of slip-induced fall accidents.

Furthermore, the onset of lower extremity fatigue during walking changed the loading rate and increased ground reaction forces (Syed and Davis, 2000) thereby reducing the forward momentum of the whole-body COM. Lockhart et al., (2003) indicated that reduced push-off force of the stance leg further reduced the transitional acceleration of the whole-body COM and increased RCOF and the risk of slip initiation. In other words, a reduction in the transitional acceleration of the whole-body COM due to localized muscle fatigue is likely to increase the friction demand at the shoe/floor interface of the contacting foot. Increased initial friction demand (i.e., RCOF) would lead to a higher likelihood of slips associated with low coefficient of friction floor surfaces. A likely factor influencing the transitional acceleration of the whole-body COM may be the ankle plantar flexors' biomechanical and physiological factors – i.e., plantar flexors produce more than half of the positive work during the push-of phase of the gait cycle (Winter, 1983). Additionally, smooth transition of the whole-body COM is maintained by hip flexion/extension to reduce the jarring effects (Inman et al., 1981). In summary, alterations of the ground reaction forces and gait kinematics due to localized muscle fatigue of the lower extremities may increase the risk of slip initiation.

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5. Aim 2: ***Effects of Fatigue on Slip Detection and Fall Recovery***

A recent review on occupational falls (Hsiao & Simeonov, 2001) summarized the major extrinsic and intrinsic factors involved in the control of balance during fall events. One of the factors that was identified as influencing risk of injury due to loss of balance is localized muscular fatigue, especially of the distal lower extremity muscles (ankle plantar flexors and knee extensors) (Sparto et al., 1997), and proximal muscles (hip and low back) (Tang et al., 1998; and Davidson, Madigan, and Nussbaum, 2004).

Nashner (1980) suggested that at the time of a potential slip-induced fall, there are certain processing stages that the central nervous system must undertake if a fall is to be avoided or recovered. During the detection of a slip perturbation, sensory input must trigger or alert those centers responsible for response selection (Lockhart et al., 2005). This alerting process may be initiated by one or more of the following sensory inputs: proprioception, vision, and vestibular function (Nashner, 1980, Mirka and Black, 1990). At the input stage, any disruption in the quality of the signal from the periphery may increase the likelihood of slips and falls. Although there is evidence of neural detectors for postural instability that could trigger the compensatory actions taken to avoid falls, their mechanism is still unclear. A study by Slobunov et al., (2005) examined the effect of postural instability on cortical activation. It was concluded that there was a burst of gamma activity (30-50 Hz brain wave emitted during high mental alertness) 200 ms prior to maximum forward lean position or the time when balance was in danger.

The literature provides support that localized muscle fatigue adversely affects proprioception (Skinner, 1986), movement coordination (Sparto, 1997), and muscle reaction times (Hakkinen, 1986), all of which may have a degrading effect on reactive recovery control of balance during slip perturbation. Possible contributions of muscle fatigue to perturbations in joint positioning sense have also been attributed to decreases in motor neuron output (Macfield, 1993, Kernell, 1982). Furthermore, Lattanzio et al., (1997) observed an impaired ability to reproduce lower extremity joint angles after fatiguing exercise. Consequently, an increase in postural sway was observed after fatigue was induced by increasing the work load (Seliga et al., 1991). Relationships between localized sensory disturbances (shoulder muscle fatigue) and global balance control were observed recently by Nussbaum (2001), and localized muscle fatigue was also found to increase postural sway. Increase in postural sway is associated with a greater risk of falling (Isaacs, 1985). As such, workers performing physically demanding tasks may compromise their ability to control balance due to disruption of the quality of the signal from sensory inputs and increase the likelihood of slip-induced falls due to delayed response selection.

Furthermore, after the response selection, the force generating capacity of lower extremity muscles may play an important role in the recovery process (attenuation of lower extremity joint motions during slip perturbation). The recovery from a fall depends largely on the strength of the lower extremity and proximal muscles which may be compromised with age and with localized muscle fatigue. In general, successful recovery from a slip event depends on substantial joint torque and power (Liu & Lockhart, 2006). Several studies have quantified the adverse effects of localized muscle fatigue on hip, knee, and ankle joint torque (Yaggie et al., 2002, Svantesson 1998, Corbeil 2003). The torque produced by these joints reduced substantially after a fatiguing protocol. The magnitude of torque generated by the contraction of muscles spanning the lower extremity joints is directly proportional to the ability to recover from an unexpected slip perturbation (Robinovitch, 2002). Studies (Do et al., 1982, Cham & Redfern, 2001) suggest that explosive strength generation and the ability to attenuate fast, large-scale lower extremity motions are critical in determining whether or not a person can respond appropriately to balance perturbation. When we experience a perturbation (from a slip), the body is set in motion, and there is a change in momentum - ultimately constrained by the generation of joint moment to reduce segmental motion and, hence, the linear momentum of the whole-body. Inability to generate the necessary counterbalancing joint moments due to fatigue during recovery either in magnitude or in rate of development to control the body's horizontal and vertical momentum can increase the risk of falls. Evidence in support of this hypothesis comes from a number of investigations indicating a decline in voluntary muscle strength, a decline in the muscle force production, and an increase in the likelihood of slips and falls. For example, Wolfson et al., (1995) and Larsson et al., (1979) reported that ankle and quadriceps muscle strength was significantly lower for those who fall as compared to non-fallers. Reduced lower extremity strength has been implicated as a factor contributing to increased risk of falling (Whipple et al., 1987). Additionally, Tang et al., (1998) indicated that proximal muscles (hip and low back) also play an important role in balance recovery. During slip perturbation, the trunk is initially extended (i.e., increases the posteriorly located position of the whole-body COM relative to the base of support) and decreases the stability requiring trunk flexion/extension to recover balance during forward slip perturbations (Oates et al., 2005). A rapid onset of an extensor/flexor synergy to position the whole-body COM anteriorly (using the proximal muscles) and downwards (using the lower limbs) has been shown to improve stability during unexpected COM displacements (Cham and Redfern, 2001). In terms of lower limb strategies to recover balance after a slip perturbation, support limb stability from the non-perturbed limb also played an important role in balance recovery and maintenance (Oddsson, et al., 2004). Furthermore, aging may affect older adults' ability to generate explosive strength even more than their ability to generate maximum strengths (Thelen et al., 1996). Since recovery of balance upon a slip perturbation requires the development of moderate-to-substantial joint moments within a short period of time (i.e., joint powers), diminished rapid torque development capacities of the older workers may require slightly longer muscular activation periods and larger activities to achieve the same mechanical effect as in the younger adults (Thelen et al., 2000). If this type of accommodation process exists, then older adults' loss of strength and execution speed may predetermine their available balance recovery strategies in the event of a slip, and may increase the likelihood of fall accidents.

Moreover, it has been theorized that muscle fatigue may impair the proprioceptive and kinesthetic properties of joints by increasing the threshold of muscle spindle discharge, disrupting afferent feedback, and subsequently altering conscious joint awareness (Balestra, 1992). The major muscles stabilizing different lower extremity joints are the rectus femoris and hamstring controlling the knee extension and flexion, whereas the gastrocnemius and tibialis anterior control ankle plantar and dorsiflexion. The effects of fatigue on the above mentioned muscles have been thoroughly investigated (Miller, 1976, Balestra, 1992). Their results demonstrated that fatigue to the knee extensors and hip flexors caused significant decreases in stabilization time compared with the fatiguing of other muscle groups. The knee joint was also identified as an important joint in terms of producing large moments while recovering from a slip (Liu and Lockhart, 2006). There is sufficient evidence that fatigue adversely affects knee proprioception (Lattanzio, 1997, Balestra, 1992). Along with the degradation of proprioception, knee and ankle stiffness is also reported to be compromised following a fatigue exertion (Wilson and Granata, 2000). These factors are directly related to the joint stability and the torque produced by the joint. It is likely that a significant portion of the injuries associated with slips and falls result from the instability of the joints as a result of fatigue of the stabilizing musculature (Troop, 1988, Lundin, 1993).

5A. Aim 2: ***Effects of Fatigue on Slip Detection and Fall Recovery***

Parijat, P., and Lockhart, T.E., (2008), Effects of lower extremity muscle fatigue on the outcomes of slip-induced falls. *Ergonomics*, 51:12, 1873-1884.

Abstract

Slip-induced fall accidents continue to be a significant cause of fatal injuries and economic losses. Identifying the risk factors causing slip-induced falls is key to developing better preventive measures to reduce fall accidents. Although epidemiolo- gical studies suggest localised muscle fatigue may be one of the risk factors for slip- induced falls, there has been no documented biomechanical study examining the relationship between fatigue and fall accidents. As such, the overall objective of the current study was to investigate the effects of localised muscle fatigue of the quadriceps on the slip initiation and slip recovery phases of slip-induced falls. Sixteen healthy, young participants were recruited to walk across a vinyl floor surface in two different sessions (fatigue and no fatigue). Kinematic and kinetic data were collected using a 3-D motion analysis system and force plates during both sessions. Results suggest that localised muscle fatigue of the quadriceps affected various kinematic and kinetic gait variables that are linked with a higher risk of slip-induced falls. Additionally, the results indicated that localised muscle fatigue of the knee extensor muscle caused a delayed response in producing an effective joint moment and base of support using the trailing limb to recover from a fall. The findings from this study indicate that localised muscle fatigue is a potential risk factor causing slip-induced falls.

Keywords: localised muscle fatigue; locomotion; fall accidents; slips and falls

Introduction

Slip and fall-related injuries and fatalities continue to pose a significant burden to industry, both in terms of human suffering and economic losses. According to the Bureau of Labor Statistics (2003), nearly 30% of workers who sustained injuries from slips and falls missed 31 or more work days. Furthermore, 14% of accidental deaths in the workplace were reportedly caused by falls (Bureau of Labor Statistics 2004). The annual direct cost of occupational injuries due to slips and falls in the US has been estimated to be in excess of $6 billion (Courtney *et al.* 2001) and is a cause of serious public health problems with costs expected to exceed $43.8 billion by the year 2020 in the US. In addition to the risk of fall-related injuries and fatalities, slip recovery efforts have been shown to contribute to high rates of overexertion injuries (Courtney and Webster 2001). It has also been documented that injuries due to falls are a major cause of years lived with disability (Murray and Lopez 1996).

According to the Bureau of Labor Statistics (2004), floors, walkways or ground surfaces were the major sources of fall accidents, causing over 86% of all fall-related injuries. Additionally, intrinsic factors, such as localised muscle fatigue (LMF), are considered as major factors contributing to slip and fall accidents (Hsiao and Simeonov 2001). Although there has been a reduction of heavy manual work attributed to growing technological advances, some occupations such as construction and forestry still demand intense physical work. The literature indicates that one-third of the US workforce exerts significant physical strength on the job and experiences fatigue in the workplace (Swaen *et al.* 2003). Previous studies have identified that changes in gait characteristics influence the risk of slip induced falls (Syed and Davis 2000, Ferber *et al.* 2002, Lockhart *et al.* 2003). Increase in friction demand characteristics and heel contact velocity (HCV) together with reduction in transitional acceleration (TA) of the whole body centre of mass (COM) during a gait cycle have been noted as risk factors for slip-induced fall accidents (Lockhart *et al.* 2003). LMF adversely affects proprioception (Skinner *et al.* 1986), movement coordination and muscle reaction times (Hakkinen and Komi 1986), which are important components of balance control. A successful recovery from a fall depends largely on the magnitude of the counterbalancing moments generated by the lower extremity joints and the rate at which these moments are generated; both of which may be compromised due to LMF.

The knee joint musculature is termed important in producing large flexion and extension moment while recovering from a slip (Cham and Redfern 2001, Liu and Lockhart 2006). The quadriceps and the hamstrings musculature aid control of knee flexion and extension and fatiguing these muscle groups may alter knee joint moment production during normal walking and recovering from a slip. Literature suggests that LMF of the quadriceps adversely affect knee proprioception and is associated with decreases in the knee joint stabilisation time (Miller *et al.* 1976, Lattanzio *et al.* 1997). Additionally, fatiguing of the lower extremity musculature may alter gait variables pertinent to slip initiation and recovery phases.

Although epidemiological studies in combination with biomechanical testing suggest LMF may be one of the risk factors for slip-induced falls, there has been no documented biomechanical study examining the relationship between the two. The overall objective of the current study was to investigate the effects of the LMF of the quadriceps on the slip initiation and slip recovery phases of slip-induced falls. The improved understanding of the relationship between LMF and slip outcome will enhance the ability to: (a) identify LMF as a potential risk factor for slip and fall accidents; (b) develop an effective intervention method (work/rest cycle schedule, exercises) to minimise the cost and rate of injury and death associated with slips and falls. It was hypothesised that LMF will adversely affect gait and recovery responses and increase slip-induced fall accidents.

Method

Participants

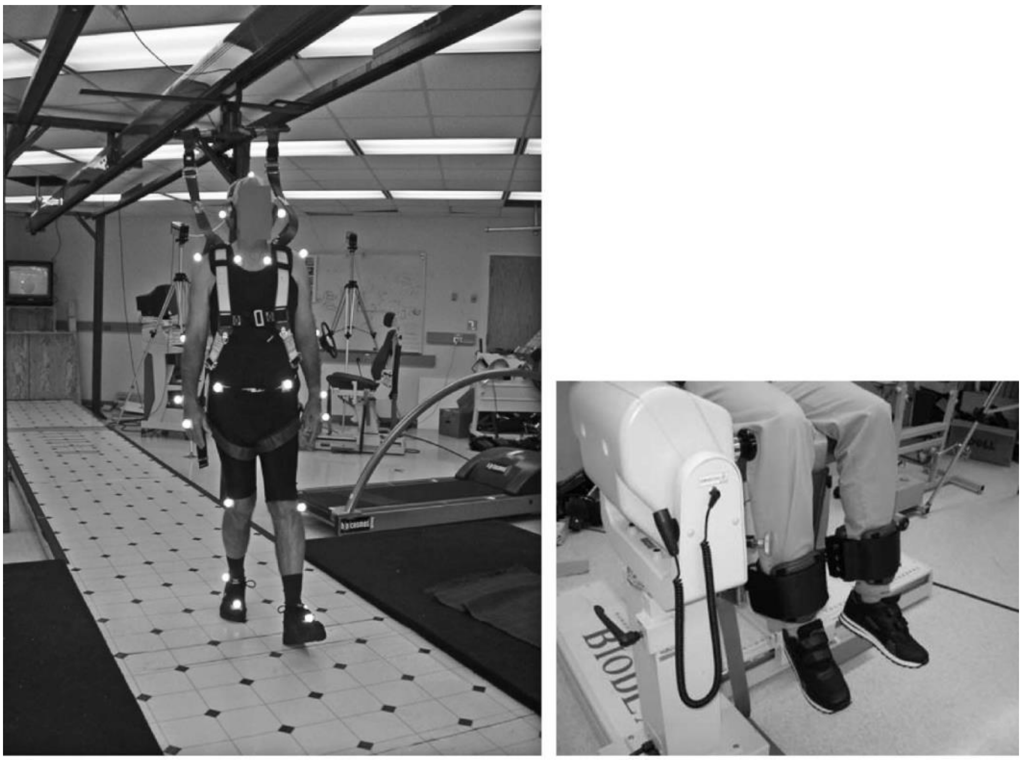
In total, 16 healthy young adults (10 males and six females) participated in the study. Informed consent was approved by the Institutional Review Board of Virginia Tech and was signed by all of the participants prior to the study. The participants (mean age 24.66 + 3.58 years, height 1.75 + 0.07 m, mass 65.86 + 10.93 kg, BMI 22.14 +2.54 kg/m^2) did not have any musculoskeletal injuries that may affect their ability to perform the fatiguing exertions.

Equipment

Walking trials were conducted on a linear walkway (1.5 $\times$ 15.5 m) embedded with two force plates (Bertec Corporation, Columbus, OH, USA). The slippery surface was covered with a water and jelly mixture (1:1) to reduce the coefficient of friction (COF) (dynamic COF was 0.12). A total of 23 reflective markers were placed over the various bony landmarks of the participants. The marker configuration of the whole body model is provided in Figure 1. A six-camera ProReflex system (Qualisys Medical AB, Gothenburg, Sweden) was used to collect 3-D position data of the participants while walking. A Biodex Dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) was used to induce fatigue. A special bilateral knee attachment was constructed for the Biodex, which essentially worked the same as with one knee attachment (Figure 1). The attachment allowed the participants to extend and flex both of their knees together. Maximum voluntary exertion (MVE) of both the knees was performed while applying minimal resistance when the joints returned to the original position. Uniform experimental shoes were provided to the participants to minimise shoe sole differences. A fall-arresting rig was used for safety (Figure 1) (Lockhart *et al.* 2003).

Fatigue inducement and experiment protocol

The experiment consisted of two different sessions, fatigue and no fatigue, within a period of 1 week. These sessions were completely randomised for all of the participants.



(a) (b)

Figure 1. (a) Reflective marker configuration of whole body along with the safety harness; (b) experiment setup for the isokinetic exertions of the knee joint using Biodex.

The randomisation process divided the sessions equally between participants. Each session consisted of normal walking and slip-perturbation trials. Bilateral quadriceps fatigue was induced using isokinetic exertions of the knees during the fatigue session. Participants were strapped on the Biodex chair with knee and hip flexed at 908. The Biodex contraption was designed to limit hip motion during the fatiguing protocol. Fatigue inducement procedures were similar to those recently described by Yaggie and McGregor (2002), with the exception that bilateral fatigue of the quadriceps was employed. The extensions were performed at 608 per s, a value consistent with earlier fatigue protocols (Kay *et al.* 2000). Participants were allowed to perform a 5- min warm-up on the Biodex and then their MVE baseline measure was recorded. After the baseline measure was recorded, participants performed bilateral knee extensions repeatedly against a resistance set at 70% of their determined baseline MVE. An MVE was performed at regular intervals (5 min) until the participants reached 60% of their baseline MVE; this was considered as the fatigue state (most of the participants reached this state within 30–50 min of exertions).

In the no fatigue session, participants were instructed to walk at a self-paced walking speed across the linear walkway for 10–15 min. Participants were asked to perform some simple tasks (filing papers) at both the ends of the linear walkway while they walked. These tasks helped to remove their attention from the floor. They were also provided with headphones to limit their hearing. The force plate and kinematic data were collected once the participants felt comfortable walking with the harness and produced consistent repetitive gait, i.e. the participant's feet landed on the centre of the force plate and in the desired sequence (right–left). Three gait cycles were recorded for each participant to represent the mean. Following this, a slippery surface was introduced without the participants' awareness and the kinetic data together with the kinematic data were collected. In the fatigue session, participants were first instructed to walk for 10–15 min and perform tasks as in the no fatigue session. After the participants produced consistent repetitive gait, they were brought to the Biodex machine for performing the fatiguing exertions. Immediately after the fatiguing protocol (5 min), the normal walking trial and slippery trial was conducted. Data were collected to represent the fatigue normal walking and slip trial. The time window of 5 min after the fatigue trial was decided from the results of a previous study (Parijat and Lockhart 2008), where it was found that participants took at least 10 min after the similar fatiguing protocol to return to their original MVE. Therefore, the time period in which the data were collected post fatigue avoided any confounding effects due to recovery from imposed muscle fatigue while walking.

Data analysis

The fatigue status (no fatigue or fatigue) was the independent variable in this study. The dependent variables consisted of various kinematic and kinetic gait variables, together with slip parameters. These parameters are divided into two groups based on the effects on slip initiation and slip recovery. The slip initiation parameters were collected during a non- slippery trial and the slip recovery parameters were collected during a slippery trial.

Gait parameters related to slip initiation

The gait parameters consisted of HCV, required COF (RCOF), TA of the whole body COM, walking velocity and peak knee joint moment. Changes in these gait parameters have been associated with the risk of slip initiation (Lockhart *et al.* 2003, Liu and Lockhart 2006). HCV was calculated by numerically differentiating the marker position data of the heel before and after the heel contact phase of the gait cycle (Lockhart *et al.* 2003). The RCOF is one of the peaks obtained from the ratio of horizontal ground reaction force to vertical ground reaction force (Fh/Fv). It represents the minimum RCOF between the shoe and floor interface to prevent initiation of forward slipping (Redfern and Andres 1984). Walking velocity was obtained from the whole body COM velocity during forward progression using the kinematic data. TA of the whole body COM was defined as the change in the horizontal COM velocity between the heel contact phase and shortly after the heel contact phase of the gait cycle (Lockhart *et al.* 2003). 2-D sagittal knee joint moment was calculated using the inverse dynamics approach (Liu and Lockhart 2006). The joint moment was normalised to body weight for data analysis. The peak value of the knee joint moment between slip start and slip recovery was utilised for the analysis.

Gait parameters related slip severity and slip recovery

The slip parameters consisted of slip distances (SDI and SDII), peak knee joint moment while recovering from slip (JMP_{slip}), joint moment activation to peak (JMAP) and timing variables (i.e.

reaction time), such as perturbed foot slip events (slip start, slip peak and slip stop) and unperturbed foot events (toe off, foot onset, foot down) were evaluated. Initial slip distance or SDI is indicative of severity of slip initiation. It is calculated from the heel marker position as the distance between the point of first minimum HCV and the point where peak heel acceleration occurs after the slip start (Lockhart *et al.* 2002). The slip distance II is indicative of the behaviour of the slip after the slip initiation (i.e. if the slip will result in a fall). The starting point for SDII is SDI slip stop and the end point is when the slip ends (i.e. first maximum of the horizontal heel velocity after the slip start) (Lockhart *et al.* 2002). It is generally accepted that a fall will occur during a slip if the slip distance exceeds 10 cm (Strandberg and Lanshammar 1981). The JMAP was defined as the time required to reach the peak joint moment while a reactive recovery attempt was made. These parameters have been used as indicators of slip severity and slip recovery in previous studies (Lockhart *et al.* 2003, Liu and Lockhart 2006).

The frequency of falls during both no fatigue and fatigue trials was considered as a dependent variable. Various parameters were utilised to detect the falls, including slip distances, sliding heel velocity and motion pictures. For a slip to be considered as a fall, the slip distance must exceed 10 cm and the peak sliding heel velocity must exceed the COM velocity while slipping (Lockhart *et al.* 2003). Additionally, videos for each of the participants were analysed to detect a fall, together with the position of the trunk marker (fall to vertical minimum).

In addition to the slip distances, timing variables were evaluated to aid in the interpretation of the slip data. For the unperturbed trailing foot, foot reaction onset (foot onset) was defined as the instant when the toe vertical position was at a maximum after toe off. Foot down was calculated as the instant when the toe vertical position was its first minimum after foot onset. The time period between foot onset and foot down (unperturbed foot reaction time) was analysed to reveal how fast the unperturbed foot could substantiate its role in the recovery process after a slip perturbation. A one-way repeated measures ANOVA was used to test for significant differences between various dependent variables during the no fatigue and fatigue sessions. The statistics were performed in the JMP 5.1 and SAS 9.1 (SAS Institute, Cary, NC, USA) statistical packages treating subject as the random effect in the ANOVA tests. The level of significance was set at a 0.05. A multivariate analysis was performed to examine the correlation between the different dependent measures.

Results

For the normal walking trials in both no fatigue and fatigue sessions, the one-way ANOVA indicated that the participants walked with a higher HCV in the fatigue session ($F(1,31)= 33.86$, $p < 0.01$) as compared to the no fatigue session (Figure 2a).The TA in the forward direction was observed to be slower during the fatigue session ($F(1, 31) = 3.85$, $p < 0.04$) as compared to the no fatigue session (Figure 2b). Consistent patterns of RCOF were observed during both of the sessions in all of the participants (Figure 3). It was also observed that RCOF was higher during the fatigue session ($F(1,31) = 9.73$, $p < 0.04$) as compared to the no fatigue session (Figure 3; Table 1).
The sagittal knee joint moment represented flexor and extensor moment in both no fatigue and fatigue sessions (Figure 4). Two distinctive extensor moment peaks (P1 and P3) were analysed for significant differences between no fatigue and fatigue sessions. Although not statistically significant ($p = 0.05$), there was a decrease in the peak extensor knee joint moment P1 in the fatigue session as compared to the no fatigue session. Peak 3, which is also extensor dominant, was decreased in the fatigue session and was significantly different from the no fatigue session ($F(1, 23) = 16.89$, $p < 0.002$).

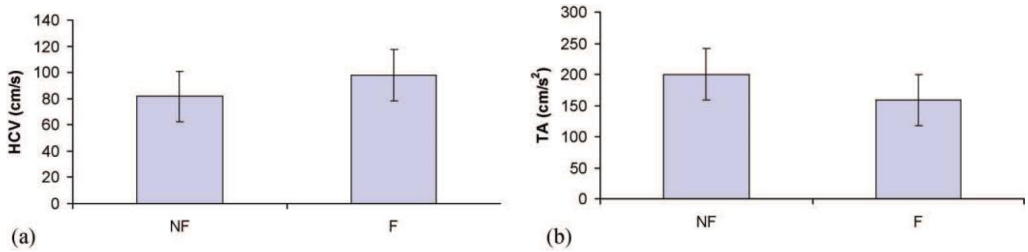


Figure 2. (a) Mean heel contact velocity (HCV (cm/s)); (b) mean (SD) transitional acceleration of the whole body centre of mass (TA (cm/s^2)) during no fatigue (NF) and fatigue (F) sessions (walking trials).

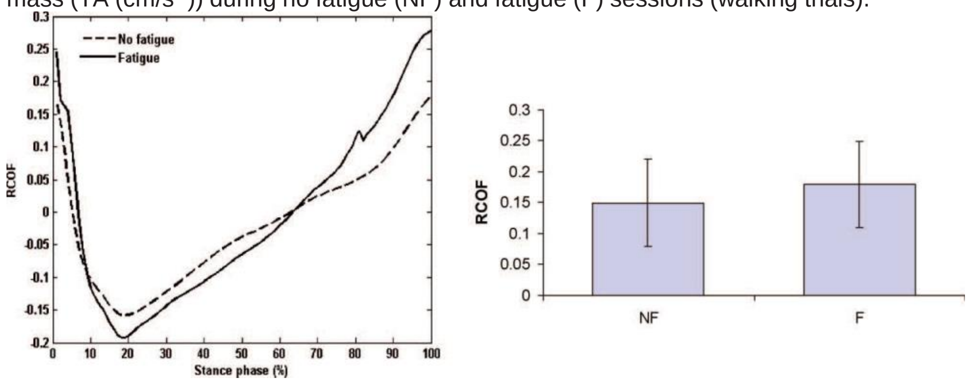


Figure 3. Ensemble average of the required coefficient of friction (RCOF) and mean (SD) RCOF during the no fatigue (NF) and fatigue (F) sessions (walking trials).

Table 1. Summary of kinematic and kinetic gait parameters.

Variables	Ses		ANOVA
	NF	F	
Slip Initiation			
HCV (cm/s)	81.94 + 51.22	97.8 + 66.67	*
WV (cm/s)	127.02 + 14.2	119.29 + 20.32	NS
TA (cm/s^2)	199.21 + 41.27	159.27 + 57.6	*
Slip Recovery			
Unperturbed foot react time (ms)	159.70 + 44.99	256.30 + 58.45	**
JMAP (ms)	333.23 + 49.8	466.48 + 88.67	*
JMP_{slip} (Nm/kg)	3.44 + 0.97	5.25 + 2.8	*
SDI (cm)	3.24 + 1.25	5.34 + 4.06	*
SDII (cm)	12.46 + 5.46	15.16 + 6.34	**

HCV = heel contact velocity; WV = walking velocity; RCOF = required coefficient of friction; TA = transitional acceleration of the whole body centre of mass; JMP_{slip} = joint moment peak during slip recovery; JMAP = joint moment activation to peak; SDI and SDII = unperturbed foot reaction time and slip distances during the no fatigue (NF) and fatigue (F) walking and slip trials.
* $p = 0.05$; ** $p = 0.01$.

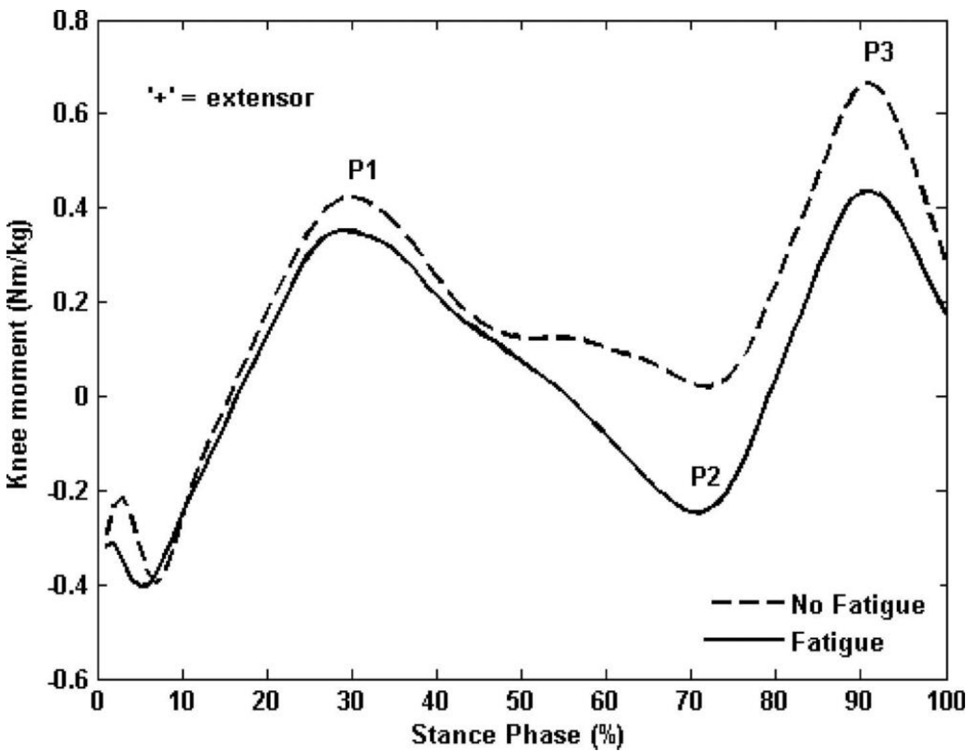


Figure 4. Average sagittal knee joint moment profile at the stance phase of gait cycle during the no fatigue and fatigue sessions (walking trials).

During the slip trials, the results indicated that participants exhibited a faster slip start in the case of a fatigue slip trial (F(1, 23) = 0.68, $p < 0.42$), but slip stop was much later as compared to the no fatigue slip trial (Figure 5). However, only the slip stop event was significantly different between the two sessions (F(1, 23) = 5.61, $p<0.03$). In terms of the unperturbed foot, the period from foot onset to foot down was faster in the case of no fatigue slip trials than fatigue slip trials (Figure 5). The knee joint moment profile for reactive recovery in both no fatigue and fatigue sessions was predominantly extensor dominant. The ANOVA performed on the recovery trials from slips revealed that the peak joint moment was higher in the fatigue slip recovery as compared to the no fatigue slip recovery (F(1,23) = 9.08, $p < 0.006$). However, the JMAP time was slower in case of fatigue slip recovery (F(1, 23) = 13.65, $p < 0.02$) (Figure 6). The slip distances SDI and SDII were both longer for slips in the fatigue session as compared to the no fatigue session (SDI, F(1,23) = 5.06, $p<0.04$, SDII, F(1,23) = 15.16, $p < 0.008$).

There were four falls in the fatigue session and one fall in the no fatigue session. The multivariate analysis between the various dependent parameters revealed correlations between certain dependent variables. The unperturbed foot reaction time and SDII (r = 0.63, $p < 0.012$) were positively correlated. Additionally, JMAP and slip stop were positively correlated (r = 0.55, $p < 0.02$), indicating that quicker reaction time leads to faster slip stop. HCV was positively correlated to RCOF (r =0.38, $p <0.02$) and SDII (r = 0.52, $p<0.012$). SDII was positively correlated to peak joint moment during recovery (r = 0.72, $p < 0.007$).

Discussion

Bilateral fatigue of the knees was employed in the current study to examine the effects of LMF on slip initiation and recovery efforts during an unexpected slip perturbation. The major findings of this study indicated that LMF of the knee alters the important gait and slip parameters that are responsible for slip initiation and recovery from a fall. The results indicated that participants walked with a higher HCV after the fatigue trials. HCV is considered important in terms of kinematic gait parameters as it can drastically change the friction demands while walking. It has been suggested that HCV affects the RCOF by altering the ratio of horizontal to vertical foot forces (Lockhart *et al.* 2003). An increase in the HCV has been considered to increase the likelihood of slip-induced falls in previous studies (Karst *et al.* 1999, Mills and Barrett 2001).

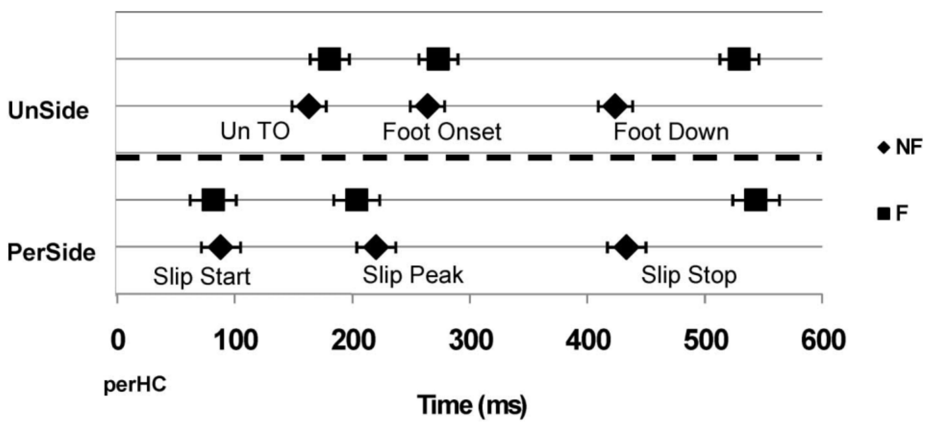


Figure 5. Occurrence of critical events after slip start during no fatigue (NF) and fatigue (F) sessions (slip trials). UnSide= unperturbed foot; PerSide =perturbed foot/slipping foot; Un TO = unperturbed foot toe off.

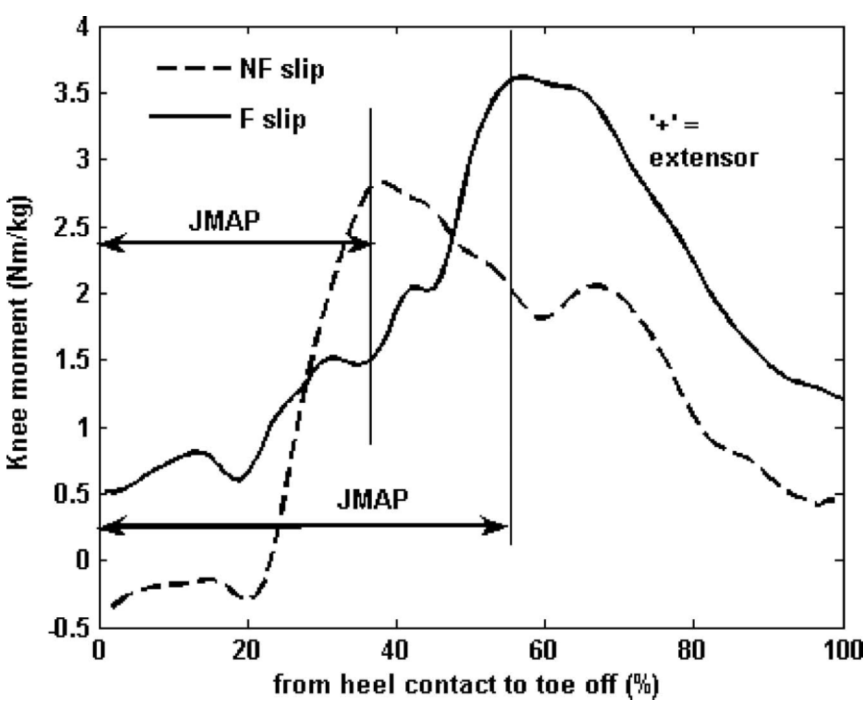


Figure 6. Average knee joint moment profile during reactive recovery in no fatigue (NF) and fatigue (F) sessions (slip trials). JMAP = joint moment activation to peak.

During an external perturbation leading to a backward fall, the speed of the forward momentum of the body is essential and an inability to produce this speed may result in a fall. The TA is an important parameter in assessing this forward momentum of the body. It was observed in this study that after the fatigue session, TA in the sagittal plane was decreased.

In terms of kinetic gait parameters during normal walking trials, the results indicated a higher friction demand during the fatigue session. The friction demand characteristics have been implicated as an important predictor variable related to severity of slips and falls (Lockhart *et al.* 2003). It has been observed that the onset of lower extremity fatigue during walking changed the loading rate and increased the ground reaction forces (Syed and Davis 2000). As RCOF is dependent on the ground reaction forces (horizontal and vertical), this would mean that increased ground reaction forces due to fatigue as observed in the current study will alter friction demand characteristics. In addition, a positive correlation was observed between HCV and RCOF, suggesting that alteration in the heel contact dynamics due to fatigue may result in increased friction demand leading to a risk of slip-induced falls. During slip trials in both the sessions, it was observed that the peak knee moment (JMP_{slip}) during the reactive recovery phase was extensor dominant and significantly higher in the fatigue slip recovery as compared to the no fatigue slip recovery. This is in agreement with the study by Ferber *et al.* (2002), which concluded a higher knee extensor moment while recovering from a perturbation. JMP_{slip} was expected to be lower in the fatigue session as LMF affects joint moment production. One of the reasons for the increase may be the recruitment of other muscle fibres (i.e. agonist co-contractions) to generate explosive strength to overcome the slip. This can also be explained based on the study by Liu and Lockhart (2006), which indicated a higher moment generation requirement while recovering from a slip. It was evident in the results that, after fatigue, participants had longer slip distances, which implied that they had severe slips. Although the magnitude of joint moment was higher, the time taken to reach the peak joint moment was slower when participants recovered after fatigue slips. Being able to rapidly develop peak joint moment was critical to balance recovery. The participants who fell were not able to rapidly produce the magnitude of joint moment required to recover from the slip. The slippery surface and all the other environmental conditions were kept constant in the study for both the fatigue and no-fatigue slip trials. However, in the fatigued state, participants slipped longer. There was a strong correlation between SDII and the peak knee moment, indicating that longer slip distance required higher knee joint moment production to recover from a fall. However, significant variability was seen in the slip distance results. This could be attributed to the complex nature of the heel contact dynamics after a slip is initiated and also to the low sample size. Further examination of these parameters is required to clearly understand the relationship between LMF and the joint moment production during a reactive recovery. Additionally, muscle activity data (electromyography) may provide some insights on how fatigued muscles respond to a perturbation.

In terms of reaction time of the unperturbed foot, foot down provided timing information on when the foot started to establish a larger base of support in order to assist an individual’s reactive recovery process. It was observed that the unperturbed foot reaction time was longer in cases of fatigue slip trials and it had a positive correlation with SDII. This led to the belief that quicker reaction of the unperturbed foot may also reduce slip distances, resulting in a faster recovery. In a study by Lockhart and Liu (2006), it was concluded that longer unperturbed foot reaction time (from onset to touchdown) may be one of the determining factors of the higher fall incidence rate for the elderly relative to their younger counterparts. Further analysis of the changes in the muscle activity due to LMF of the quadriceps and alterations in the sensory input may help in clearly defining a relationship between LMF and the foot reaction time during reactive recovery.

Based on the discussion above, a descriptive model was created to indicate the changes in the dependent factors due to LMF during phases of slips and falls. Figure 7 illustrates the relationship between LMF and the different dependent variables together with their correlation found in the current study. It should, however, be noted that further biomechanical analyses and a larger sample size is required to completely understand these mechanisms. Additionally, many covariates are not considered in this illustration. The increase in friction demand characteristics and HCV have been linked with increased risk of slip initiation (Lockhart *et al.* 2003). The current study concluded that LMF adversely affects HCV and RCOF. Additionally, the current study revealed that there was an increase in peak knee joint moment and unperturbed foot reaction times in the fatigue slip trials. These factors were correlated to SDII. All of these variables have been shown to affect the slip recovery phase leading to falls (Liu and Lockhart 2006, Lockhart and Liu 2006). Thus, LMF may affect the slip recovery phase by altering these parameters. The solid lines in Figure 7 indicate the alterations in the dependent measures due to LMF found in the current study, which are related to slip initiation and recovery phases.

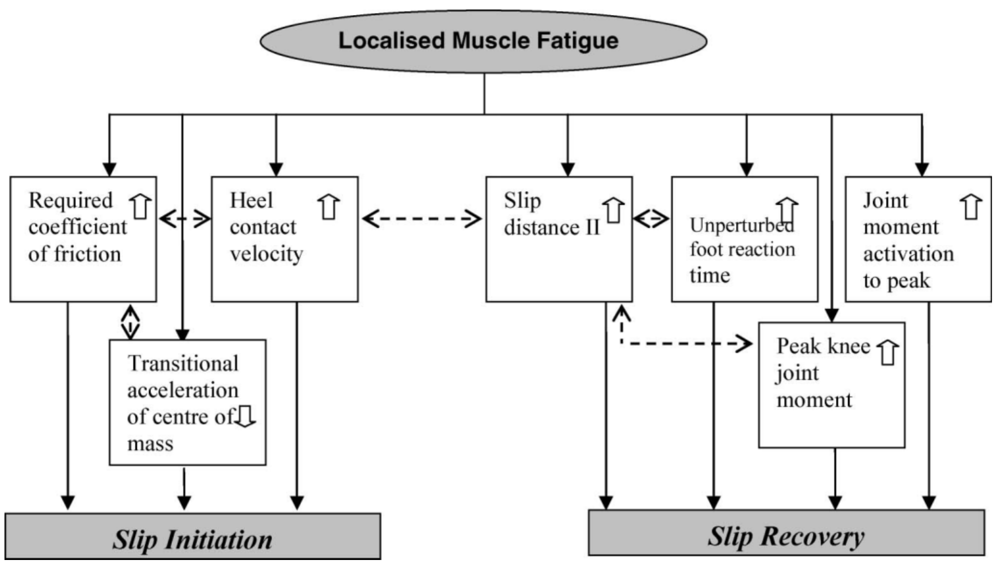


Figure 7. Flowchart of all the factors associated with slip initiation and slip recovery that were significantly different after fatigue. Solid line arrows indicate the changes observed; dotted line arrows indicate the correlation between the variables.

The dotted lines represent the correlation between the variables. Although implicated, the correlations between the different gait and slip parameters are complex and this might challenge the interpretation of the results. In addition, the variability in the data may be responsible for masking certain characteristics of LMF related to slip-induced falls. Further biomechanical analysis of the parameters is required to understand the relationship between LMF and phases of slip-induced falls.

Conclusion

In summary, LMF of the quadriceps affects various kinematic and kinetic gait variables that are linked with a higher risk of slip-induced falls and, therefore, can be considered as a potential risk factor for slip-induced falls. Additionally, the results also indicate that LMF of the knee extensors caused a delayed response in producing joint moment and increasing the base of support using the trailing limb. One of the limitations of the study was that each participant reached their fatigue level at a different time. These limitations can affect the results due to the difference in the fatigue level of each individual. Additionally, quadriceps musculature consists of four different muscle groups (rectus femoris, Vastus lateralis, Vastus intermedius and rectus femoris). The current study did not isolate muscle fatigue to each of the muscles but as the quadriceps muscle group as a whole. There might be limitations as rectus femoris muscle does not fatigue in the same way as other muscles in this group. Although implicated, the 60% of baseline MVE as a fatigue state prior to testing ensured that all participants were fatigued at similar levels.

Future research will investigate the effects of LMF of multi-joint fatigue (i.e. hamstrings, ankle plantar flexors) on slip events in a real-world job scenario. However, results from the present study can be used as preliminary information on the specific gait and slip parameters that are sensitive to LMF. Other potential areas for further research include evaluating the effects of rest breaks and recovery and effects of age on fatigue.

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5B. Aim 2: ***Effects of Fatigue on Slip Detection and Fall Recovery***

Ehsan, R., Bochen, J., Nussbaum, M., and Lockhart, T., (2013). Investigating the effects of slipping on lumbar muscle activity, kinematics, and kinetics. 2013 International Annual Meeting of the Human Factors and Ergonomics Society, San Diego, California, September 30-October 4, 2013.

Abstract
Slips, trips, and falls remain leading causes of occupational injuries and fatalities. The current exploratory study sought to quantify lumbar kinematics and kinetics during both induced slips and normal walking. Individual anthropometry, lumbar muscle geometry, and lumbar kinematics, along with electromyography of 14 lumbar muscles were used as input to a 3D, dynamic, EMG-based model of the lumbar spine. Results indicate that, in comparison with values during normal walking, lumbar kinematics, lumbosacral kinetics, lumbar muscle activations, and lumbosacral reaction forces were all substantially increased during a slip event. Observed levels of muscle activity and lumbosacral reaction forces suggest the potential for low back injury during a slip event. Outcomes of this work may facilitate the identification and control of specific mechanisms involved with low back disorders consequent to a slip.

INTRODUCTION
Slips, trips, and falls (STF) continue to be significant occupational safety issues (Kemmlert & Lundholm, 2001; T. B. Leamon & P. L. Murphy, 1995) and important causes of occupational injuries and fatalities (Bureau of Labor Statistics, 2011; T. Leamon & P. Murphy, 1995). Several factors such as physical hazards and surface contamination can contribute to STF prevalence. Uneven or contaminated surfaces, stairs and ladders, inclined surfaces, and improper shoe soles are among identified risk factors (Bentley & Haslam, 2001; Grönqvist et al., 2001). Additional risk factors are related to working postures, work speeds, and task demands (Redfern et al., 2001).

Among STF events, slipping is a common and typically unpredictable event often leading to falls on the same level or to a lower level (Andersson & Lagerlöf, 1983; Courtney, Sorock, Manning, Collins, & Holbein-Jenny, 2001). More specifically, floor slipperiness may contribute to 40 - 50% of fall-related injuries (Courtney et al., 2001). Depending on the extent of a given slip event, there can be a need for a balance recovery effort to prevent a fall. These recovery efforts, again dependent on the magnitude of the slip and other potential factors, can be anticipated to be substantial in some case due to the sudden movements and unexpected excessive loads on body parts. Of particular interest here are loads on the low back during a slip-recovery event (Stobbe & Plummer, 1988). Although previous studies have reported back pain as an outcome of slip and trip incidents (Grönqvist et al., 2001; Manning & Shannon, 1981; Pope, 1989), existing biomechanical analyses of slip events have focused primarily on the lower extremity (Cham & Redfern, 2001). In contrast, there is no quantitative evidence available, to the authors' knowledge, regarding spinal loads. Therefore, the objective of the current exploratory study was to quantify lumbar muscle activity, kinematics, and kinetics (i.e., spine loading) during a slip.

METHODS
Participants

Six participants (5M, 1F) were recruited from the local community and completed informed consent procedures approved by the Virginia Tech Institutional Review Board prior to their participation in the study. Participation was limited to individuals free of any self-reported musculoskeletal disorders or injuries during the past 2 years and who were currently physically active. Participants' mean (SD) age, stature, and body mass were 27.0 (1.7) yrs, 179.0 (11.1) cm, and 73.2 (11.3) kg, respectively.

Experimental Design and Procedures

A repeated-measures design was used, in which participants completed several walking trials and one trial involving an unexpected slip. General procedures for these were similar to methods used in several previous studies (Beschorner & Cham, 2008; Liu & Lockhart, 2009), and so are only briefly summarized here. Initial measurements were made of resting muscle activity, followed by several maximal exertions of the trunk and placement of reflective surface markers (see below). After several walking trials at a natural cadence, a slippery surface was unexpectedly introduced. A force platform was embedded underneath the floor surface for both conditions. To ensure safety, during all trials participants wore a fall arrest harness that was connected to an overhead rail. To minimize variability due to shoe sole properties, a consistent type of shoe was used by participants during the experiment.

Data Collection

Muscle activity (EMG: electromyography) was measured during resting, maximal exertion, and walking trials. EMG was obtained bilaterally from several muscles in the lower lumbar region, including three flexors [internal oblique (IO), external oblique (EO), rectus abdominis (RA)] and four extensors [iliocostalis lumborum pars lumborum (ILL), multifidus (MF), longissimus thoracis pars lumborum (LTL), and longissimus thoracis pars thoracis (LTT)]. Bipolar Ag/AgCl electrodes (AccuSensor, Lynn Medical, MI) with a 2.5 cm inter-electrode distance were placed as described previously (Jia, Kim, & Nussbaum, 2011). Raw EMG signals were collected using two telemetered systems (TeleMyo 900, Noraxon, AZ, USA), sampled at 800 Hz, band-pass filtered (20-400 Hz), rectified, low-pass filtered at 5 Hz (bidirectional 1st-order Butterworth), and normalized using individual maximal and minimal signal values.

Maximal voluntary muscle activation was measured from three sets of maximal voluntary contractions (MVCs) including trunk flexion/extension, clockwise and counterclockwise axial rotation, and left/right lateral bending. Participants were asked to reach their maximum level of muscle activity with fast and jerky exertions. In contrast to more traditional ramped efforts, these procedures were used to facilitate model calibration for the dynamic movements involved in recovering from an induced slip. During MVCs, participants stood in a customized fixture attached to a force platform (AMTI OR6-7-2000, Watertown, MA, USA), and movements were restricted using knee, pelvis, and shoulder straps (Jia et al., 2011). All exertions were performed in an upright posture except the extension trials, which were done in 20 degree of trunk flexion. Resting values of muscle activation were recorded in quiet prone and supine positions.

Segmental kinematics were obtained using a 6-camera optoelectronic system (ProReflex, Qualysis, Gothenburg, Sweden), which was used to track the locations of 25 passive retro-reflective markers placed over bony landmarks of the foot, shank, thigh, pelvis, trunk, and head (Damsgaard, Rasmussen, Christensen, Surma, & de Zee, 2006; Davis, Tyburski, & Gage, 1991). Kinematic data were recorded at 100 Hz, then low pass filtered with a cutoff frequency of 5 Hz (bidirectional 1st-order Butterworth). Finally, ground reaction forces were sampled (800 Hz) from two identical force platforms (BERTEC # K80102, Type 45550-08, Columbus, OH, USA), and low-pass filtered at 12 Hz (bidirectional 1st-order Butterworth).

Biomechanical Models

Lumbar muscle geometry and kinematics were estimated using an anatomical model developed in the AnyBody™ musculoskeletal modeling system (v5.0, AnyBody Technology, Aalborg, Denmark). Initial insertions, via points, and origins of a total of 76 muscle fascicles were adopted from pre-defined values in the AnyBody repository, and were scaled based on individual anthropometry. The AnyBody model was "driven" using collected marker data, and provided output that included lumbar kinematics (torso rotations relative to the pelvis) and the lengths, velocities, and moment arms of the muscle fascicles. These outputs, along with normalized EMG (nEMG) and individual anthropometry, were used as input to a 3D, dynamic, EMG-based model of the lumbar spine (Jia et al., 2011). Lumbar muscle forces, along with lumbosacral (L5/S1) reaction moments and internal forces (compression and shear forces) were calculated as a function of: nEMG; nEMG-to-force relationship; muscle stress (maximal force per unit area); physiological cross-sectional area; passive tissue contribution; and functional properties (electromechanical delay; force-length and force-velocity relationships). The noted EMG-based model has been described in more detail elsewhere (Jia et al., 2011) and evaluated for a range of task demands.

Analysis

Outcome measures were compared between normal walking and slip trials using one-way, repeated measures analyses of variance (ANOVAs). Statistical analyses were performed using JMP 9.0 (SAS Institute Inc., Cary, NC, USA), and significance was determined when $p < 0.05$. All summary data are presented as means (SD).

RESULTS

Overall, lumbar kinematics, L5/S1 moments, lumbar muscle activity, and L5/S1 forces were all substantially increased during slip events versus during normal walking. For brevity, representative results from one participant are presented blow, for a slip event involving the right leg and from which the participant recovered (i.e., not caught by the safety harness). Subsequently, summary results are given for L5/S1 loads.

Lumbar Kinematics and L5/S1 Moments

Flexion/extension motion of the lumbar spine was small during the initial phase of the example slip event (Figure 1). Note, though, that the figure indicates relative rotations of the torso vs. the pelvis. Later in the event (i.e., ~0.5 s following heel contact), there was a substantial extension motion (Figure 1). The range of flexion/extension movement for the slipping trial was 18 degrees versus 8 degrees in normal walking for this participant. At right heel contact, left lateral flexion was observed (contralateral side), and the range of lateral bending for the slipping trial was 23 degrees vs. 13 degrees for normal walking. Axial rotation of the torso relative to the pelvis was toward the side of heel contact, and respective ranges of axial rotation were 21 and 13 degrees for the slipping trial and normal walking.

Lateral bending moments at L5/S1 oscillated around zero during normal walking, and heel contact of the right foot was concurrent with the generation of a right-directed lateral bending moment (0-0.4 sec following heel contact). The maximum lateral bending moment (Figure 2) was almost four times higher in the slipping trial versus the normal walking one. The maximum flexion/extension moment occurred approximately at toe off during normal walking. In the slip trial, the maximum flexion moment increased 5-fold in comparison to normal walking. While walking, axial twist moments were smaller relative to the other components, with maximum values at approximately toe off. At each step, and in the stride interval from toe off to the following heel contact, a twist moment was generated in the contralateral side. Nearly a three-fold increase in peak axial moment was observed during the slip trial.

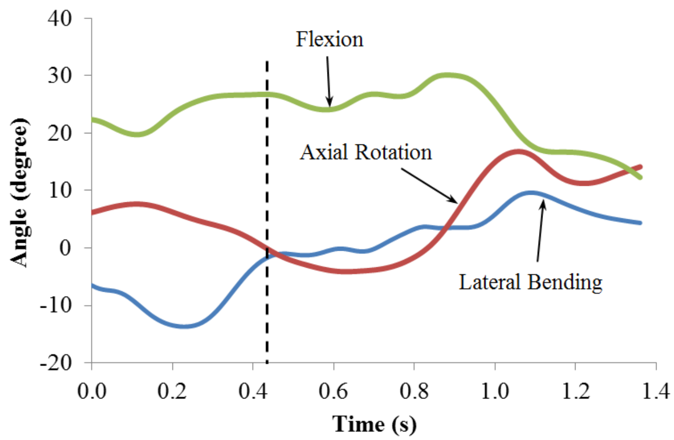

Figure 1. Lumbar kinematics (torso relative to pelvis). Data are shown for one slipping trial, with right foot heel contact at 0.43 second (dashed black line). Positive values indicate flexion, right lateral bending, and twisting to the right.

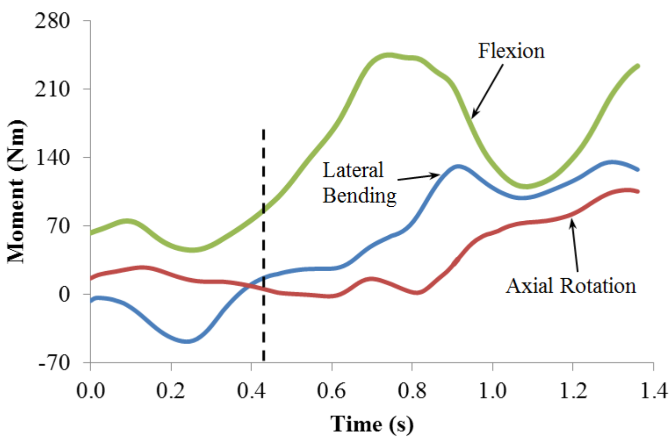

Figure 2. Triaxial L5/S1 reaction moments for a slip trial, with right foot heel contact at 0.43 second (dashed black line). Positive values indicate flexion, right lateral bending, and twisting to the right.

Lumbar Muscle EMG

nEMG for the bilateral lumbar muscles are depicted for a slip trial in Figure 3. All muscles demonstrated a substantial increase in activation very soon after right heel strike (0.43 seconds), and reached maximum activation levels after roughly 0.4 sec. All of the extensor muscles activated with a similar pattern during the slip event. Left-side flexors activated out of phase with the extensors, presumably to bring the left leg forward faster, and exhibited the highest values just before left heel strike. Generally, the level of muscle activation was much lower for normal walking in comparison to the slip trial (Table 1). Maximum values of muscle activation were seven times higher for the slipping trial versus the normal walking on average (Table 1). Of note, several muscles approached full activation during the slip event (e.g., 100% of MVC), with similar results for other participants.

Table 1. Maximum values of bilateral normalized lumbar muscle activations during normal walking and a slipping trial.

Trial		Muscle activation (% MVC)						
		MF	LTL	ILL	LTT	EO	IO	RA
Normal Walking	Right	9	21	13	18	11	9	6
	Left	14	14	13	9	15	7	4
Slip	Right	100	100	100	80	73	91	100
	Left	88	81	79	43	76	97	100

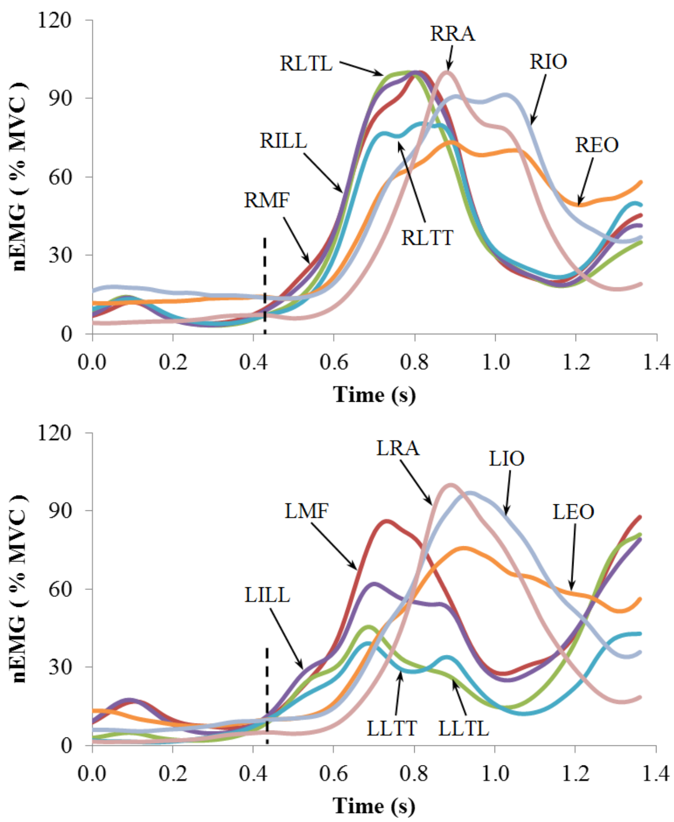

Figure 3. Normalized EMG (nEMG) of the lumbar muscles in a slipping trial on the right (top) and left (bottom) sides. Right foot heel contact was at 0.43 second (dashed black line).

Lumbosacral Forces

As expected, L5/S1 axial forces remained entirely compressive throughout both trials (Figure 4). During normal walking, the triaxial forces exhibited two peaks, approximately between the time of heel strike for one leg and toe off for the other one. All three forces were substantially larger for this participant during the trip event. Across all participants, there were large increases in the maximum lumbosacral shear and compressive forces in the slip vs. normal walking trials (Table 2).

DISCUSSION

Although previous studies have investigated lumbar kinematics (Murray, Mollinger, Gardner, & Sepic, 1984; Rowe & White, 1996), muscle activity (Cappozzo, 1984; Murray et al., 1984), and low back loads (Cappozzo, 1983, 1984; Rowe & White, 1996) during gait, the current study provides what appears to be a novel detailed examination of lumbar kinematics and kinetics during the more dynamic conditions present during a slip event. Normal walking imposed low to moderate loads on the spine. However, slipping increased lumbar motions and muscle activations, and introduced significant lumbosacral reaction moments and forces.

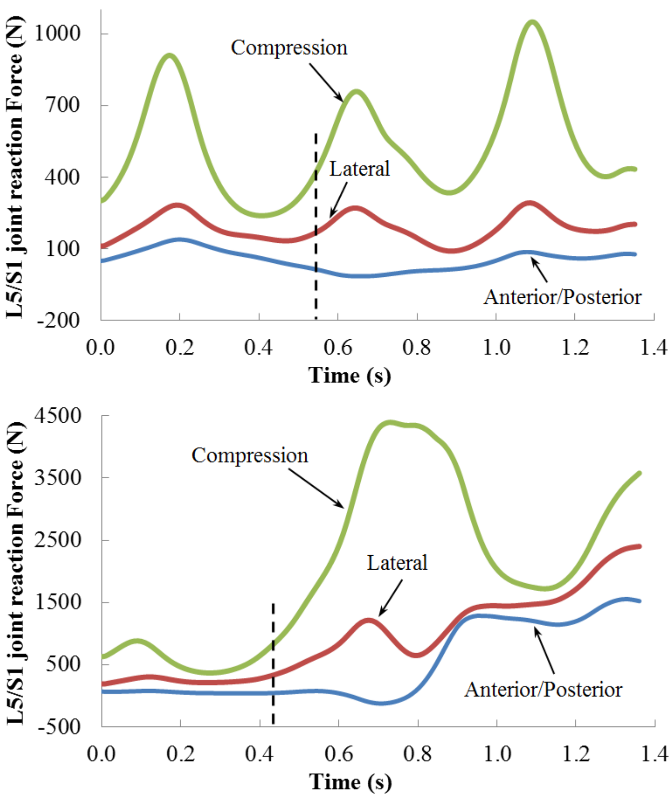

Figure 4. Joint reaction forces at L5/S1 level for one participant including anterior-posterior and lateral shear, and compression force for normal walking (top) and a slip trial (bottom). Right foot heel contacts occurred at 0.54 and 0.43 seconds, respectively (dashed black lines)

Table 2: Maximum lumbosacral reaction forces, predicted using the EMG-driven model. Values are means (SDs) across participants, and *p* values indicate differences between walking and slipping trials.

Trial	Anterior/posterior shear (N)	Lateral shear (N)	Compression (N)
Normal Walking	102 (40)	204 (49)	1003 (194)

Slip	791 (565)	1167 (796)	4728 (1124)
p value	0.042	0.026	0.0002

The current EMG-based model used measured kinematics and incorporated actual muscle activation level to predict lumbar muscle forces and lumbosacral reaction forces. More investigation is needed to validate the results of the current study, and is currently being undertaken by comparing predicted moments with values estimated from inverse dynamics analyses.

The current outcomes predicted during normal walking, however, are in reasonable agreement with previous work. Two earlier studies using a single muscle model (Cappozzo, 1984; McGill, 1992) predicted compressive forces that were 145-207% and 100-250% of body weight (BW), respectively, during normal walking. A similar study (Khoo, Goh, & Bose, 1995), using an EMG-driven model, found peak compressive forces of between 92 and 345% of BW. Predicted peak compressive forces during normal walking in the current study were 141(39) % of BW across participants. Regarding anterior/posterior lumbosacral shear forces, during gait Khoo et al. (1995) reported a mean value equal to 22% BW, while Callaghan & McGill (2001) estimated it to be 15%. Anterior/posterior shear forces in the present study were estimated to be 15(6) % of BW across participants. Moreover, lateral shear was estimated as 12-58% of BW by Callaghan et al. (1999), similar to the values of 30(10) % predicted here. Of note, the observed patterns of lumbar kinematics here during normal walking were also qualitatively and quantitatively consistent with earlier reports (Cheng, Chen, Chen, & Lee, 1998; Rowe & White, 1996).

Previous studies have highlighted back pain as a potential outcome of slip and trip incidents (Grönqvist et al., 2001; Manning & Shannon, 1981; Pope, 1989). More generally, spinal loads (compression and shear) are among the likely important biomechanical mechanisms involved in LBD onset, especially in occupational environments. It is notable that in all slip events here the peak compression force exceeded the NIOSH (1981) “Action Limit” of 3400 N and all peak resultant shear forces exceeded a proposed 500 N Action Limit (McGill, Norman, Yingling, Wells, & Neumann, 1998). Further, the initial phases of a slip event involved very high levels of muscle activity. These high levels of activity across all muscles (*cf.* Table 1), including substantial levels of antagonistic co-contraction, may represent a reflexive “stiffening” of the trunk in the early phases of a slip event. Such high levels of activation, though, could also contribute to musculotendinous damage. The present results thus suggest that, in additional to causes subsequent to a slip (e.g., contact with the ground from a fall), low-back injury could result from high levels of forces generated during the slip event itself. Outcomes of this study may help to identify potential slip-related mechanisms in the development of LBDs as well as future preventive approaches.

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6. Supplemental Studies

i. Liu, J., and Lockhart, T.E., (2009), Age-related joint moment characteristics during normal gait and successful reactive recovery from unexpected slip perturbations. *Gait and Posture,* 30, 276-281.

Abstract

The objective of the current study was to investigate the effects of aging on 3D lower extremity joint moments during successful reactive-recovery from unexpected slips. Unexpected slips were induced by having participants walk over a slippery floor surface. Successful reactive-recovery trials from nine young and nine elderly participants were identified and analyzed. Three-dimensional inverse dynamics were implemented to calculate reactive joint moments at the ankle, knee, and hip joints. Peak joint moment magnitude and the speed of peak joint moment generation were used to describe the balance recovery strategies from unexpected slips. Results indicated significantly higher peak joint moments in recovery than in normal walking for both the young and elderly. Meanwhile, during reactive-recovery, the elderly were found to utilize both frontal and sagittal joint moments while the younger adults relied primarily on sagittal joint moment. It was concluded that the ankle and knee joints were critical in controlling sagittal plane motion disturbance, while the hip joint was mainly responsible for stabilizing upper body balance in the frontal plane. This study confirmed age-related differences in joint moment generation during unexpected slips. Additionally, implementing 3D analysis is recommended in future slips and falls research. PMID:19581088 [PubMed - indexed for MEDLINE] PMCID:PMC3716287

ii. Kim, S., Lockhart, T.E., and Nam, C.S., (2010), Leg Strength Comparison between younger and middle-age adults. *International Journal of Industrial Ergonomics*, 40: 315-320.

Abstract

Although a risk of occupational musculoskeletal diseases has been identified with age-related strength degradation, strength measures from working group are somewhat sparse. This is especially true for the lower extremity strength measures in dynamic conditions (i.e., isokinetic). The objective of this study was to quantify the lower extremity muscle strength characteristics of three age groups (young, middle, and the elderly). Total of 42 subjects participated in the study: 14 subjects for each age group. A commercial dynamometer was used to evaluate isokinetic and isometric strength at ankle and knee joints. 2 × 2 (Age group (younger, middle-age, and older adult groups) × Gender (male and female)) between-subject design and Post-hoc analysis were performed to evaluate strength differences among three age groups. Post-hoc analysis indicated that, overall, middle-age workers' leg strengths (i.e. ankle and knee muscles) were significantly different from younger adults while middle-age workers' leg strengths were virtually identical to older adults' leg strengths. These results suggested that, overall, 14 middle-age workers in the present study could be at a higher risk of musculoskeletal injuries. Future studies looking at the likelihood of musculoskeletal injuries at different work places and from different working postures at various age levels should be required to validate the current findings. The future study would be a

valuable asset in finding intervention strategies such that middle-age workers could stay healthier longer.PMID:20436934[PubMed] PMCID:PMC2861367

iii. Parijat, P., and Lockhart, T.E., (2012), Effects of moveable platform training in preventing slip-induced falls in older adults. *Annals of Biomedical Engineering*, 40(5):1111-21.

Abstract

Identifying effective interventions is vital in preventing slip-induced fall accidents in older adults. The purpose of the current study was to evaluate the efficacy of moveable platform training in improving recovery reactions and reducing fall frequency in older adults. Twenty-four older adults were recruited and randomly assigned to two groups (training and control). Both groups underwent three sessions including baseline slip, training, and transfer of training on a slippery surface. Both groups experienced two slips on a slippery surface, one during the baseline and the other (after 2 weeks) during the transfer of training session. In the training session, the training group underwent twelve simulated slips using a moveable platform while the control group performed normal walking trials. Kinematic, kinetic, and EMG data were collected during all the sessions. Results indicated a reduced incidence of falls in the training group during the transfer of training trial as compared to the control group. The training group was able to transfer proactive and reactive control strategies learned during training to the second slip trial. The proactive adjustments include increased center-of-mass velocity and transitional acceleration after training. Reactive adjustments include reduction in muscle onset and time to peak activations of knee flexors and ankle plantar flexors, reduced ankle and knee coactivation, reduced slip displacement, and reduced time to peak knee flexion, trunk flexion, and hip flexion velocities. In general, the results indicated a beneficial effect of perturbation training in reducing slip severity and recovery kinematics in healthy older adults.PMID:22134467[PubMed - indexed for MEDLINE] PMCID:PMC3319506

iv. Wu, X., Lockhart, T.E., & Yeoh, H., (2012), Effects of obesity on slip-induced fall risks among young male adults. *Journal of Biomechanics, 45;6: 1042-1047.*

Abstract

Obesity is associated with structural and functional limitations with impairment of normal gait. Although falls have been identified as the most common cause of injuries in the obese, the mechanisms associated with increased fall risk among the obese population are still unknown. The purpose of this study was to investigate the influence of gait adaptations of the obese individuals and its implication on risk of slip initiations as measured by friction demand characteristics. To exclude the aging and gender effects, a total of ten healthy young male adults participated in the study. Kinematic and kinetic data were collected using a three-dimensional motion analysis system and force plates while subjects were walking at their self-selected walking pace. Results indicated that young obese adults walked similarly as their lean counterparts except for exhibiting greater step width and higher transversal friction demand, suggesting that slip-induced fall risks are similar along the horizontal direction, but increased along the transversal direction under certain floor conditions.PMID:22304846[PubMed - indexed for MEDLINE] PMCID:PMC3310324

v. Liu, J., Zhang, X., and Lockhart, T.E., (2012), Fall risk assessments based on postural and dynamic stability using inertial measurement unit. *Safety and Health at Work*, 3: 192-198.

Abstract

Slip and fall accidents in the workplace are one of the top causes of work related fatalities and injuries. Previous studies have indicated that fall risk was related to postural and dynamic stability. However, the usage of this theoretical relationship was limited by laboratory based measuring instruments. The current study proposed a new method for stability assessment by use of inertial measurement units (IMUs). Accelerations at different body parts were recorded by the IMUs. Postural and local dynamic stability was assessed from these measures and compared with that computed from the traditional method. THE RESULTS DEMONSTRATED: 1) significant differences between fall prone and healthy groups in IMU assessed dynamic stability; and 2) better power of discrimination with multi stability index assessed by IMUs. The findings can be utilized in the design of a portable screening or monitoring tool for fall risk assessment in various industrial settings.PMID:23019531[PubMed] PMCID:PMC3443694

vi. Liu, J., and Lockhart, T.E., (2013), Local dynamic stability changes associated with load carrying. *Safety and Health at Work*, 4,1: 46-51.

Abstract

Load carrying tasks are recognized as one of the primary occupational factors leading to slip and fall injuries. Nevertheless, the mechanisms associated with load carrying and walking stability remain illusive. The objective of the current study was to apply local dynamic stability measure in walking while carrying a load, and to investigate the possible adaptive gait stability changes. Current study involved 25 young adults in a biomechanics research laboratory. One tri-axial accelerometer was used to measure three-dimensional low back acceleration during continuous treadmill walking. Local dynamic stability was quantified by the maximum Lyapunov exponent (maxLE) from a nonlinear dynamics approach. Long term maxLE was found to be significant higher under load condition than no-load condition in all three reference axes, indicating the declined local dynamic stability associated with load carrying. Current study confirmed the sensitivity of local dynamic stability measure in load carrying situation. It was concluded that load carrying tasks were associated with declined local dynamic stability, which may result in increased risk of fall accident. This finding has implications in preventing fall accidents associated with occupational load carrying.PMID:23515183[PubMed] PMCID:PMC3601296

vii. Fino, P. and Lockhart, T.E., (2014), Required coefficient of friction during turning at self-selected slow, normal, and fast walking speeds. *Journal of Biomechanics,* 47(6): 1395-1400. PMID: 24581815

Abstract

This study investigated the relationship of required coefficient of friction to gait speed, obstacle height, and turning strategy as participants walked around obstacles of various heights. Ten healthy, young adults performed 90° turns around corner pylons of four different heights at their self selected normal, slow, and fast walking speeds using both step and spin turning strategies. Kinetic data was captured using force plates. Results showed peak required coefficient of friction (RCOF) at push off increased with increased speed (slow μ=0.38, normal μ=0.45, and fast μ=0.54). Obstacle height had no effect on RCOF values. The average peak RCOF for fast turning exceeded the OSHA safety guideline for static COF of μ>0.50, suggesting further research is needed into the minimum static COF to prevent slips and falls, especially around corners. PMID:24581815[PubMed - indexed for MEDLINE] PMCID:PMC4054705

viii. Liu, J., & Lockhart, T. E. (2014). Trunk angular kinematics during slip-induced backward falls and activities of daily living. Journal of Biomechanical Engineering, 136(10), 101005. PMID: 25033029

Abstract

Prior to developing any specific fall detection algorithm, it is critical to distinguish the unique motion features associated with fall accidents. The current study aimed to investigate the upper trunk angular kinematics during slip-induced backward falls and activities of daily living (ADLs). Ten healthy older adults (age = 75 ± 6 yr (mean ± SD)) were involved in a laboratory study. Sagittal trunk angular kinematics were measured using optical motion analysis system during normal walking, slip-induced backward falls, lying down, bending over, and various types of sitting down (SN). Trunk angular phase-plane plots were generated to reveal the motion features of falls. It was found that backward falls were characterized by a simultaneous occurrence of a slight trunk extension and an extremely high trunk extension velocity (peak average = 139.7 deg/s), as compared to ADLs (peak average = 84.1 deg/s). It was concluded that the trunk extension angular kinematics of falls were clearly distinguishable from those of ADLs from the perspective of angular phase-plane plot. Such motion features can be utilized in future studies to develop a new prior-to-impact fall detection algorithm. PMID:25033029[PubMed - in process] PMCID:PMC4127473

ix. Liu, J., Lockhart, T. E., and Kim, S., (2014), Reaction moment at the L5/S1 joint during a simulated forward slipping with a handheld load, *International Journal of Occupational Safety and Ergonomics , 20(3): 429-436.*

Abstract

The purpose of the study was to investigate the effects of load on the net moment response at the L5/S1 joint during simulated slip events. Six young individuals were instructed to take one step with a handheld load. Sudden floor movement was randomly introduced to simulate unexpected slips. Different loads conditions (0%, 10%, 20%, 30% of body weight) were introduced at random. Three-dimensional net moments at the L5/S1 joint were computed via downward inverse dynamic model. Peak joint moment generated at 30% load level was found to be significantly higher compared to no-load condition. No peak moment differences were found among no-load, 10% or 20% load levels. Additionally, the findings from this study indicated a flexiondominant net L5/S1 joint moment pattern during motion phase associated with slip-induced falls. PMID:25189747

x. Liu, J., & Lockhart, T.E. (2014) Development and evaluation of a prior-to-impact fall event detection algorithm, *IEEE Transactions on Biomedical Engineering*, 61(7): 2135-2140. PMID: 24718566

Abstract

Automatic fall event detection has attracted research attention recently for its potential application in fall alarming system and wearable fall injury prevention system. Nevertheless, existing fall detection research is facing various limitations. The current study aimed to develop and validate a new fall detection algorithm using 2-D information (i.e., trunk angular velocity and trunk angle). Ten healthy elderly were involved in a laboratory study. Sagittal trunk angular kinematics was measured using inertial measurement unit during slip-induced backward falls and a variety of daily activities. The new algorithm was, on average, able to detect backward falls prior to impact, with 100% sensitivity, 95.65% specificity, and 255 ms response time. Therefore, it was concluded that the new fall detection algorithm was able to effectively detect falls during motion for the elderly population. PMID:24718566

LIST OF PUBLICATIONS

1. Parijat, P., Lockhart, T.E., and Liu, J., (in press), Perturbation-based slip training using a virtual reality environment on slip-induced falls. *Annals of Biomedical Engineering*. DOI: 10.1007/s10439-014-1128-z. PMID: 25245221

2. Parijat, P., Lockhart, T.E., and Liu, J., (in press), EMG and kinematic responses to unexpected slips after slip training in virtual reality. *IEEE Transactions in Biomedical Engineering.* DOI: 10.1109/TBME.2014.2361324. PMID: 25296401

3. Fino, P. and Lockhart, T.E., (2014), Required coefficient of friction during turning at self-selected slow, normal, and fast walking speeds. *Journal of Biomechanics,* 47(6): 1395-1400. PMID: 24581815

4. Liu, J., & Lockhart, T.E. (2014) Development and evaluation of a prior-to-impact fall event detection algorithm, *IEEE Transactions on Biomedical Engineering*, 61(7): 2135-2140. PMID: 24718566

5. Liu, J., & Lockhart, T. E. (2014). Trunk angular kinematics during slip-induced backward falls and activities of daily living. Journal of Biomechanical Engineering, 136(10), 101005. PMID: 25033029

6. Zhang, J., Lockhart, T.E., and Soangra, R., (2014), Classifying Lower Extremity Muscle Fatigue during Walking using Machine Learning and Inertial Sensors. *Annals of Biomedical Engineering,* 42(3): 600-612. PMID: 24081829

7. Soangra, R., Lockhart, T.E., Lach, J., and Abdel-Rahman, E., (2013), Effects of hemodialysis therapy on sit-to-walk characteristics in End Stage Renal Disease patients. *Annals of Biomedical Engineering*, 41,4:795-805. PMID: 23212801

8. Liu, J., and Lockhart, T.E., (2013), Local dynamic stability changes associated with load carrying. *Safety and Health at Work*, 4,1: 46-51. PMID: 23515183

9. Lockhart, T.E., and Stergiou, N., (2013), New Perspectives in Human Movement Variability. *Annals of Biomedical Engineering*, Vol 41:8, 1593-1594. PMID: 23797778

10. Wu, X., Lockhart, T.E., & Yeoh, H., (2012), Effects of obesity on slip-induced fall risks among young male adults. *Journal of Biomechanics, 45;6: 1042-1047.* PMID: 22304846

11. Haynes, C., Lockhart, T.E., (2012), Evaluation of gait and slip parameters for adults with intellectual disability. *Journal of Biomechanics.* 45: 2337-2341. PMID: 22867766

12. Parijat, P., and Lockhart, T.E., (2012), Effects of moveable platform training in preventing slip-induced falls in older adults. *Annals of Biomedical Engineering*, 40(5):1111-21. PMID: 22134467

13. Liu, J., Zhang, X., and Lockhart, T.E., (2012), Fall risk assessments based on postural and dynamic stability using inertial measurement unit. *Safety and Health at Work*, 3: 192-198. PMID: 23019531

14. Kim, S., and Lockhart, T.E., (2012), Lower Limb Control and Mobility Following Exercise Training. *Journal of Neuroengineering and Rehabilitation*, 15;9(1). PMID: 22335998

15. Park, S.H., Kim, S., Lockhart, T.E., Yoon, H.Y., and Lee, K., (2011), Effects of Sound on Postural Stability During Standing. *Journal of Neuroengineering and Rehabilitation*, 8:67. PMID: 22168248

16. Liu, J., and Lockhart, T.E., (2013), Automatic individual calibration in fall detection: an integrative ambulatory measurement framework. *Computer Methods in Biomechanics and Biomedical Engineering*, 16(5):504-10. PMID: 22149355

17. Lockhart T.E., Barth A, Zhang X, Songra R, Abdel-Rahman E, Lach J. (2010), Portable, non-invasive fall risk assessment in end stage renal disease patients on hemodialysis, *ACM Trans Comput Hum Interact.* 2010: 84-93. PMID: 22124286

18. Kim, S., and Lockhart, T.E., (2010), Effects of 8 weeks of balance or weight training for the independently living elderly on the outcomes of induced slips. *International Journal of Rehabilitation Research*, 33(1): 49-55. PMID: 19773670

19. Liu, J., and Lockhart, T.E., (2009), Age-related joint moment characteristics during normal gait and successful reactive recovery from unexpected slip perturbations. *Gait and Posture,* 30, 276-281. PMID: 19581088

20. Lockhart, T.E., Kim, S., Kapur, R., and Jarrott, S., (2009), Evaluation of gait characteristics and ground reaction forces of cognitively declined older adults with emphasis on slip-induced fall accidents. *Assistive Technology, 21(4):188-195.* PMID: 20066885

21. Kim S., and Lockhart, T.E., Roberto, K., (2009), The effects of 8-week balance training or weight training for the elderly on fear of falling measures and social activity levels. *Quality in Ageing*, 10(4), 37-48. PMID: 21394234

22. Lee, M., Roan, M., Smith, B., and Lockhart, T.E., (2009), Gait analysis to classify external load conditions using linear discriminant analysis. *Human Movement Science*, 28, 226-235. PMID: 19162355

Conference

1. Soangra, R., Lockhart, T. E. (2014). Understanding relationship between center of pressure signals from forceplates and accelerometers. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting, 2014*

2. Chung, C., Soangra R., Lockhart, T. E. (2014). Recurrent Quantitative Analysis of Postural Sway using Forceplate and Smartphone. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting, 2014*

3. Chang W., Leclercq, S., Haslam, R., and Lockhart, T., (2013). The state of Science on Occupational Slips, Trips and Falls on the Same Level. The Proceedings of the International Conference on Fall Prevention and Protection, National Institute of Occupational Safety and Health, Japan (JNIOSH), Tokyo, pp.33-40. (Keynote Address).

4. Frames C, Soangra R, Lockhart TE. (2013). Assessment of postural stability using inertial measurement unit on inclined surfaces in healthy adults - biomed 2013. *Biomed Sci Instrum*. 49:234-42.

5. Lockhart TE, Soangra R, Zhang J, Wu X. (2013). Wavelet based automated postural event detection and activity classification with single IMU. *Biomed Sci Instrum*. 49:224-33.

6. Soangra R, Lockhart TE. (2013). Comparison of intra individual physiological sway complexity from force plate and inertial measurement unit. *Biomed Sci Instrum*. 49:180-6.

7. Wu, X., and Lockhart, T.E., (2013). Fall risk assessment for the community dwelling obese elderly using wearable sensors. 37th Annual Meeting of the American Society of Biomechanics, Omaha, Nebraska, September 4-7, 2013.

8. Soangra, R., Lockhart, T.E., and Zhang, J., (2013). Effects of Electromyogram Signal filtering on Muscle Activation Time. 37th Annual Meeting of the American Society of Biomechanics, Omaha, Nebraska, September 4-7, 2013.

9. Zhang, J., Lockhart, T.E., and Soangra R., (2013). Ankle Fatigue Classification Using Support Vector Machines. 37th Annual Meeting of the American Society of Biomechanics, Omaha, Nebraska, September 4-7, 2013.

10. Wu, X., and Lockhart, T.E., (2013). Fall Risks Assessment and Fall Prediction among Community Dwelling Elderly Using Wearable Wireless Sensors. 2013 International Annual Meeting of the Human Factors and Ergonomics Society, San Diego, California, September 30-October 4, 2013.

11. Wu, X., Yeoh, H., and Lockhart, T.E., (2013). 10-meter walking test among obese and non-obese community dwelling elderly The XXVth Annual Occupational Ergonomics and Safety Conference, Atlanta, GA, June 6-7, 2013.

12. Lockhart, TE, Yeoh, HT, Soangra, R., Jongprasithporn, M., Zhang, J., Wu, X., and Ghosh, A. (2012). Non-invasive fall risk assessment in community dwelling elderly with wireless inertial measurement units. *Biomed Sci Instrum*, 48:260-7. PMID: 22846292

13. Soangra, R., Lockhart, TE. (2012). Determination of stabilogram diffusion analysis coefficients and invariant density analysis parameters to understand postural stability associated with standing on anti-fatigue mats. *Biomed Sci Instrum,* 48:415-22. PMID: 22846314

14. Seimetz, C., Tan, D., Katayama, R., and Lockhart, TE. (2012). A Comparison Between Methods Of Measuring Postrual Stability: Force Plates Versus Accelerometers. *Biomed Sci Instrum,* 48:386-92. PMID: 22846310

15. Soangra, R., and Lockhart, TE. (2012). A comparative study for performance evaluation of sit-to-stand task with body worn sensor and existing laboratory methods. *Biomed Sci Instrum,* 48:407-14. PMID: 22846313

16. Soangra R, Lockhart TE and Van de Berge N. (2011)**.** An approach for identifying gait events using wavelet denoising technique and single wireless IMU, *Proceedings of the Human Factors and Ergonomics Society Annual Meeting, 55: 1990-1994.*

17. Soangra R, and Lockhart TE. (2011). An Approach for Identifying Posturo-Locomotion-Manual Events Using Wavelet Denoising Technique and Three Wireless IMU, *Proceedings of Biomedical Engineering Society* (BMES 2011), PS – Fri – A – 94.

18. Lockhart TE., Yeoh HT, Soangra R, Haynes C., Brolinson PG. (2011). Effects of Orthotics on Dynamic Stability in the Elderly, *Proceedings of Biomedical Engineering Society* (BMES 2011) PS – Fri – A – 187.

19. Jongprasithporn, M., Lockhart, T.E., and Yodpijit, N. (2011). The Age-Related Effect on Slip Severity and Response Time During Slip Perturbations, Proceedings of the XXIII Annual International Occupational Ergonomics and Safety Conference, Baltimore, Maryland, USA, June 9-10, 2011.

20. Liu, J., Kim, S., and Lockhart, T.E. (2011). *Classification of daily activities for the elderly using wearable sensors*. Human Computer Interaction International 2011, Orlando, FL.

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22. Zhang, X., and Lockhart, T. E. (2010). The impairment and recovery of dynamic walking stability during virtual environment exposure in the elderly. *Paper accepted by the Human Factors and Ergonomics Society 54th Annual Meeting*, San Francisco, CA.

23. Zhang, X., & Lockhart, T. E. (2010). Locomotion stability adaptation to the virtual reality induced sensory conflicts. T*he 16th US National Congress of Theoretical and Applied Mechanics*, State College, PA.

24. Yodpijit, N., and Lockhart, T.E. (2010). "Investigating Dynamics of Dark Focus of the Human Eye" The 54th Annual Meeting of the Human Factors and Ergonomics Society, September 27-October 1, San Francisco, CA.

25. Yodpijit, N., and Lockhart, T.E. (2010). "Stability of Dark Focus of the Human Eye" the XXIInd Annual International Occupational Ergonomics and Safety Conference, June 10-11, Tempe, Arizona.

26. Zhang, X., Lockhart T. E. (2009). A reliability study of three functional mobility assessment tools in fall risk evaluation. *Proceedings of the Human Factors and Ergonomics Society 53rd Annual Meeting,* 1719-1723, San Antonio, TX.

27. Zhang, X., Lockhart T. E., and Parijat, P. (2009). Fall risk assessment based on Postural-Locomotor-Manual test utilizing miniature accelerometers. *Proceedings of the International Ergonomics Association 17th World Congress on Ergonomics,* Beijing, China.

28. Yodpijit, N., Jongprasithporn, M., and Lockhart, T.E. (2009). “Revised Horizontal Working Area for Southern Thai Population” Proceedings of the XXIst Annual International Occupational Ergonomics and Safety Conference, Dallas, Texas, USA, June 11-12, p. 142-144.

29. Jongprasithporn, M., Yodpijit, N., and Lockhart, T.E. (2009). “Slips, Trips, and Falls Training for Delivery Workers” Proceedings to the XXIst Annual International Occupational Ergonomics and Safety Conference 2009, Dallas, Texas, USA, June 11-12, p. 163-166.

30. Zhang, X., Lockhart, T. E., and Liu, J. (2009). The use of inertial measurement units in assessing standing and walking stability for fall risk estimation. *Proceedings of the International Society for Occupational Ergonomics and Safety 21st Annual Conference,* 150-156, Dallas, TX.

31. Liu, J., & Lockhart, T. E. (2009). A novel integrative ambulatory measurement framework - application in fall event detection*. Proceedings of the XXIst Annual International Occupational Ergonomics and Safety Conference*, 157-162, Dallas, TX.

32. Liu, J., & Lockhart, T. E. (2009). Trunk angular kinematics during slip-induced falls and activities of daily living - towards developing a fall detector*. Proceedings of the Human Factors and Ergonomics Society 53rd Annual Meeting*, 892-896, San Antonio, TX.

33. Parijat, P., Haynes, C.A., Lockhart, T.E., and Antin, J., (2009), Investigation of biomechanical characteristics of older adults: Effects of gender and driving status. *Proceedings of the Human Factors and Ergonomics Society 53rd Annual Meeting,* 1724-1727, San Antonio, TX.

34. Haynes, C.A., and Lockhart, T.E., (2009), Differences in gait parameters between non-disabled and intellectually disabled adults. *Proceedings of the Human Factors and Ergonomics Society 53rd Annual Meeting,* 1719-1723, San Antonio, TX.

Ph.D. Students

1. Prakriti Parijat (2009), *Trainability of Recovery Responses in Older Adults to Prevent Slip-Induced Falls using Simulated Slip Training.*
2. Natakrit Yodpijit (2010), *The Effect of Age on Dark Focus Distance and Visual Information Transfer Rate.*
3. Manutchanok Jongprasithporn (2011), *The Age-Related Effects of Visual Input on Multi-Sensory Weighting Process During Locomotion and Unexpected Slips.*
4. Selina Zhang (2013), *Determinants of Gait and Fall Risk*.
5. Rahul Soangra (2014), Understanding variability in older adults using inertial sensors.
6. Jian Zhang (2014), Machine learning and prediction of fallers using wearable sensors.

INCLUSION OF GENDER AND MINORITY STUDY SUBJECTS

Study Title: Effects of Localized Muscle Fatigue on Risks of Occupational Slips and Falls
Total Enrollment:
Grant Number: R01 OH009222

PART A. TOTAL ENROLLMENT REPORT: Number of Subjects Enrolled to Date (Cumulative) by Ethnicity and Race				
	Sex/Gender			
Ethnic Category	**Females**	**Males**	**Unknown or Not Reported**	**Total**
Hispanic or Latino	3	1	0	4 **
Not Hispanic or Latino	27	29	0	56
Unknown (individuals not reporting ethnicity)	0	0	0	0
Ethnic Category: Total of All Subjects*	30	30	0	60 *
Racial Categories				
American Indian/Alaska Native	0	0	0	0
Asian	1	2	0	3
Native Hawaiian or Other Pacific Islander	0	0	0	0
Black or African American	2	2	0	4
White	23	24	0	47
More Than One Race	1	1	0	2
Unknown or Not Reported	0	0	0	0
Racial Categories: Total of All Subjects*	27	29	0	56 *

PART B. HISPANIC ENROLLMENT REPORT: Number of Hispanics or Latinos Enrolled to Date (Cumulative)				
Racial Categories	**Females**	**Males**	**Unknown or Not Reported**	**Total**
American Indian or Alaska Native	0	0	0	0
Asian	0	0	0	0
Native Hawaiian or Other Pacific Islander	0	0	0	0
Black or African American	0	0	0	0
White	1	1	0	2
More Than One Race	0	0	0	0
Unknown or Not Reported	0	0	0	0
Racial Categories: Total of Hispanics or Latinos**	3	1	0	4 **

INCLUSION OF CHILDREN

Most of the studies conducted under this grant involved children (i.e. those 18-21 years of age). Many workers are children by this standard, and are exposed to conditions of interest here, specifically where there prolonged and/or repetitive lifting. Hence, the results of the research described here have relevance for conditions affecting children.

MATERIALS AVAILABLE FOR OTHER INVESTIGATORS

All experimental data are available. Software (MatLab™ code) is available for the biomechanical models. These may be accessed by contacting the PI, whose information is provided below. Note that the investigators wish to limit such access to individuals requesting data or software for research purposes only.