Effects of Quadriceps Fatigue on the Outcomes of Slips and Falls

Prakriti Parijat

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Graduate Committee Members:

Thurmon E. Lockhart, Ph.D.,Committee Chair
Kevin Granata, Ph.D., Committee Member
John Cotton, Ph.D., Committee Member

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ABSTRACT

Identifying potential risk factors that affect slip-induced falls is key to developing effective interventions for reduction of injuries caused by these accidents. 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 literature on the relationship between localized muscular fatigue of the lower extremity and the gait parameters affecting slip severity is scarce. The purpose of the present study was to examine how lower extremity fatigue (quadriceps) alters gait parameters and increases slip severity. Sixteen healthy young participants were recruited to walk across an unexpected slippery floor 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 gait parameters important in assessing slip severity were compared for the two different sessions to evaluate the effects of fatigue. A repeated measure one-way analysis of variance (ANOVA) and multivariate analysis was employed to predict statistical significance. The results indicated a substantial increase in the heel contact velocity (HCV), required coefficient of friction (RCOF), slip distance II (SDII), peak average knee joint moment during slip recovery ($\text{kneemom}_{\text{peak}}$), fall frequency and, a decrease in the transitional acceleration of the whole body COM (TA) in the fatigue session further indicating higher slip severity due to fatigue. In addition, a strong positive correlation was observed between RCOF and HCV, HCV and SDII, and, SDII and $\text{kneemom}_{\text{peak}}$. These findings provide new insights into the relationship between localized muscular fatigue and slip initiation/recovery process. The present study concluded that localized muscular fatigue affects the gait parameters and increases slip severity and hence can be considered as a potential risk factor for slip-induced falls.

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ABBREVIATIONS

ANOVA	Analysis of variance
COM	Centre of mass
F	Fatigue session
HCV	Heel contact velocity
$\text{Kneemom}_{\text{peak}}$ (slip)	Peak knee joint moment of the slipping foot
$\text{Kneemom}_{\text{peak}}$(norm)	Peak knee joint moment during normal stance phase while walking.
MVE	Maximum voluntary exertion
NF	No fatigue session
PSHV	Peak sliding heel velocity
RCOF	Required coefficient of friction
RPE	Rating of perceived exertion
SDI	Slip distance I
SDII	Slip distance II
TA	Transitional acceleration of the whole body centre of mass
WV	walking velocity or whole body centre of mass velocity

Chapter 1

To be conscious that we are perceiving or thinking is to be conscious of our own existence…Aristotle

1.0. Introduction

Slips and falls related injuries and fatalities continue to pose a significant burden to industry as well as health sector, 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 days of work or more. Further, 14% of accidental deaths in 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 US dollars (Courtney et al., 2001), and a cause of serious public health problem with costs expected to exceed $43.8 billion by the year 2020 in the U.S. 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 (Murrey 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 occupationally induced localized muscle fatigue, are considered as major factors contributing to slip and fall accidents (Bentley and Haslam, 2001; Cohen and Lin, 1991; Maiti et al., 2001,; Gauchard et al., 2001; Hsiao and Simeonov, 2001). 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). Localized muscular fatigue has been identified as a potential risk factor for various musculoskeletal injuries and has been successfully applied to reduce disorders caused by them (i.e., recommended lifting capacity). In the context of fall accidents, the existing evidence (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 and control during slip perturbation and increase the risk of slips and falls.

Although there has been a reduction of heavy work in industrialized countries due to growing technology, labor demanding intense physical work is still necessary in some occupations such as construction, forestry and many service occupations (Astrand and Rodahl, 1986). Literature indicates 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). Though epidemiological studies suggest localized muscular fatigue might be one of the risk factors for slip-induced falls, there has been no documented biomechanical study examining the relationship between the two. However, in the past few decades, research has been more focused in understanding the biomechanics involved in the process of slips and falls/recovery. For example, Ferber et al. (2002) investigated the effect of unexpected forward perturbations during gait (walking) on lower extremity (ankle, knee & hip) joint mechanics. Some distinct lower extremity joint moment and power patterns which prevented collapse during the perturbations were observed in their study. Lockhart et al. (2003) examined the different gait characteristics (step length, heel contact velocity, coefficient of friction) between the young and the elderly to quantify the frequency of falls in the elderly. These studies have helped us understand that change in gait characteristics is directly related to slip propensity and also to the corrective action taken by the body in response to a perturbation.

Recently, lower extremity fatigue has been linked to slip and fall accidents. This was based on several studies which have shown effects of localized muscle fatigue on gait characteristics. As discussed above, changes in gait will have a significant effect on the slip propensity and also on recovery from a slip. In a recent assessment on occupational falls by Hsiao and Simeonov (2001), the major extrinsic and intrinsic factors involved in the control of balance during the slip-induced fall events were summarized. One of the intrinsic factors that have been identified to influence balance is the localized muscular fatigue, especially of the lower extremity muscles (ankle plantar flexors and knee extensors). The lower extremities are the prime movers during a gait cycle and thus fatigue of this musculature might have detrimental effects on the gait characteristics which in turn might increase the slip propensity.

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 mechanism associated with muscle fatigue and its relationship with slip-induced fall accidents still remains unclear. A complete understanding of biomechanics of slips and falls and the effects of localized muscular fatigue is required for successful fall interventions. The following section of this thesis will explain in detail the effects of fatigue on gait characteristics along with the biomechanics of slips and falls.

1.1. Background and Significance

1.1.1. Biomechanics of Gait during slips and falls

The successful intervention solutions for reducing slips and falls require complete understanding of the mechanism involved. A review of literature on slip and fall accidents indicates that multiple mechanisms are involved in slip and fall accidents. In general, the ability to walk safely and preserve balance in the event of an external perturbation (slip) is dependent upon intact visual, vestibular, proprioceptive, and musculoskeletal systems. However, with factors like advancing age and muscular fatigue, a variety of physiological changes may interfere with gait and balance, placing individuals at a high risk for slip and fall accidents.

Traditionally, the analysis of normal walking consists of a complete gait cycle which occurs from heel strike of one foot to the consecutive heel strike of that same foot. The cycle is divided into two separate phases: 1) stance phase and 2) swing phase. The stance phase accounts for approximately 60% of one cycle while swing phase accounts for the remaining 40% (Perry, 1992). Biomechanical analysis usually involves kinematic and kinetic analyses. Kinematics, by definition, is the description of human movement independent of forces that cause movement (Winter, 1991). They include linear and angular displacements, velocities, and accelerations. Numerous methods of recording gait parameters are available: kinematic (motion analysis) systems, electromyography, measurements of energy cost, and force platform systems (Begg et al., 1989). Kinetics, by definition is the study of body motion with respect to forces and energy (Winter, 1991). One of the most valuable kinetic variables associated with the assessment of human movement is the time history of the lower extremity joint moments (Winter, 1991). The net joint moment is defined as the product of net muscle forces times the moment arm measured from the joint center to segment center (Winter, 1991). Inverse dynamics is the most common approach utilized to determine the lower extremity joint moments. The Table 1.1 summarizes all the important gait parameters utilized in slips and fall studies.

Table 1.1 Various Gait variables utilized in biomechanical analysis of slips and falls

Gait parameters	
Kinematics	***Kinetic***
Stride length, Peak sliding (HCV)	**Ground reaction forces - Horizontal (Fh), Vertical (Fy)**
Heel Contact Velocity (HCV)	**RCOF (Fh/Fv)**
Centre of mass (COM) velocity	**Joint moment**
Slip Distance I (Between slip start and mid-slip)	**Joint work**
Slip Distance II (Between mid-slip and Slip-end)	**Joint power**
Displacements, velocity and acceleration	**Muscle Force**

To understand the biomechanics of slips and falls, it is important to first understand the different phases involved during this process. The process of slips and falls is categorized into four levels as shown in figure 1.1-1. The environmental phase considers the effects of contamination. As noted by Chaffin et al. (1992), "any fluid contaminant between two sliding surfaces will provide lubrication and thereby lower the dynamic coefficient of friction (DCOF) values". Therefore, presence of contamination (oil, water, etc.) will reduce the available DCOF of the floor surfaces. In terms of slip-induced falls, friction demand characteristics between the shoe sole surface and the floor surface has been implicated as an important predictor variable related to severity of falls. It was stated that most of slip-induced falls occurred when the frictional force (F_μ) opposing the direction of foot movement is less than the shear force (F_h) of the foot immediately after the heel contact on the floor (Perkins and Wilson, 1983).

The required coefficient of friction has also been utilized to study the potential for slip-induced falls between age groups (Lockhart, 2000). The 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 (Perkins, 1978). Consequently, slip is initiated by the combination of low DCOF and higher RCOF. A static COF of 0.5 on level walking surface has been commonly recommended by standard organizations and by individual authors (Lin et al., 1995). Dangerous forward slips that lead to falls are most likely to occur 70-120 ms after the heel contacts the ground (Gronqvist, 1989; Perkins, 1978). The number of slip and fall events increases as the difference between the RCOF and available dynamic COF of the floor surface increases (Bonder, 1994). Thus, changes in RCOF as a function of localized muscle fatigue can provide useful insights into the risk of slip initiation.

Initial gait characteristics such as stride length and heel velocity may affect RCOF because of the increase in horizontal foot force. An investigation of older individual's gait characteristics by Lockhart et al. (2000) revealed that the risk of slip-induced falls in this age-group may be due to higher heel velocity which affects their RCOF. It has also been suggested that whole body centre of mass (COM) velocity relative to the base of support may be a factor related to RCOF (Lockhart et al., 2003; Pai et al., 1997). Slower whole body COM velocity and COM transitional acceleration (velocity changes from heel contact to shortly after heel contact) were reported in older adults (Lockhart et al., 2003).

During the detection and recovery phases of the slips and falls process, the central nervous system (CNS) control plays an important role. The CNS must undertake certain processing stages (detection phase) if a fall is to be avoided or compensated for (recovery phase) (Lockhart et al., 2006). During the detection phase, a trigger must be sent through the sensory feedback to the motor control regions of the CNS. This process may be initiated by one or more of the following sensory inputs: proprioception, vision, and vestibular function. At the input stage any disruption in the quality of the signal from the periphery may increase the likelihood of slips and falls. Though there is an 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), which is the brain wave emitted during high mental alertness 200 ms prior to maximum forward lean position or the time when balance was in danger. Furthermore as shown in Figure 1.1, the recovery from a fall also depends largely on the strength of the lower extremity which might be compromised with age and with factors like localized muscular fatigue.

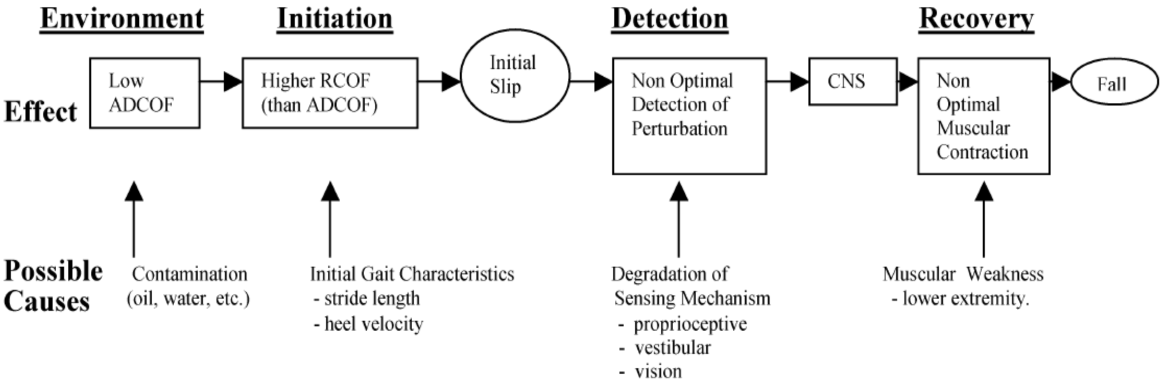


Figure1.1. The process of initiation, detection, and recovery of inadvertent slips and falls with possible causes and effects (adopted Lockhart et al, 2003)

1.1.2. Localized muscular fatigue and its physiological effects

Localized muscle fatigue is defined as the inability of the muscles to maintain expected force output (Vollestad, 1997). Fatigue is a very complex phenomenon which is widely known but very 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 muscles' demands for oxygen supply and lactate removal which results in reduced endurance, 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 decrease in the maximum force producing ability.

Fatigue can also be broadly divided into peripheral and central level. At the peripheral level, it causes a failure in the transmission of the neural signal or a failure of the muscle to respond to neural excitation. Peripheral fatigue is more of a local fatigue that affects one muscle or a muscle group. The factors contributing to peripheral fatigue include metabolic inhibition of the contractile process and excitation- contraction coupling failure (Miller et al., 1995; Baker et al., 1993; Cady et al., 1989). At the central level, fatigue may cause failure of excitation of the motor neuron (Corbeil et al., 2003). Central fatigue is critical to physical performance since the lack of adequate CNS drive to the working muscles result in fatigue in most people during normal activities (Davis, 1996). The origin of fatigue (central or peripheral) is critical to the understanding of fatigue.

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 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 to 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.

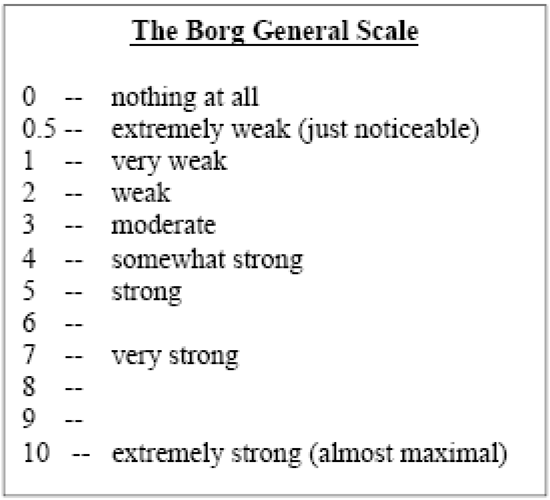
The onset of localized muscular fatigue is detrimental to productivity and performance of workers due to alterations in various physiological and biomechanical characteristics. The accumulations of metabolic by products reduce the intra muscular conduction velocity until a point where the muscles are unable to produce the desired force output (Gefen et al., 2001). Literature provides support that localized muscle fatigue adversely affects proprioception (Skinner, 1986), movement co-ordinations (Sparto, 1997), and muscle reaction times (Hakkinen and Komi, 1986). All these factors as discussed in section 1.1, affect balance as well. Possible contributions of muscle fatigue to perturbations in joint positioning sense have been attributed to decrease in motor neuron output (Macfield et al., 1993; Kernell et al., 1982). Further, Latanzio et al. (1997) observed an impaired ability to reproduce lower extremity joint angles after fatiguing exercise. 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 like ion shifts, electrical excitation and shifting concentrations of substrates and metabolites also occur. These changes are conveyed to the central nervous system, causing sensations of effort and discomfort (perceived fatigue).

1.1.2.1. Fatigue Measurement

Fatigue can be measured subjectively through the use of scales or questionnaires or objectively through physiological methods such as electromyography. Some of the rarely used objective measures include changes in posture, altered muscle coordination, and changes in performance accuracy. There has been a 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 decrease in maximal ability, it can be measured directly by taking maximum voluntary contraction (MVE) measurements over the course of dynamic activity. The value of the participant's MVE is considered as a "gold standard" for identification of fatigue occurrence (Vollestad, 1997). MVE is defined as the maximum force generated when the participant is encouraged to perform a contraction to their highest ability.

The use of EMG is an indirect measure as EMG does not directly measure maximum ability. Also, during dynamic activities more than one muscle group is involved in producing the required force and thus it becomes very difficult to isolate a muscle or muscle group to measure its activity. However, it cannot be said with certainty that the direct measurements (MVEs) are more valid because even MVEs can be affected by individual factors such as motivation. Many studies have used the decrease in peak torque as a measure of assessment of the development of fatigue in isokinetic muscle test. According to Gerdle and Langstrom (1987), fatigue should be assessed by the decrease in work.

The Borg Rating of Perceived Exertion (RPE) Scale (Borg, 1985) is a formal technique which has been used to subjectively assess fatigue in human subjects. This scale provides a very simple rating method to measure the perceived exertion in which the values grow fairly linear with respect to the work load (Borg, 1985). Over the years, the RPE scale has been modified to the Borg General Scale shown in Figure 1.2. It is a linear scale that ranges from 0 to 10 where 0 represents no fatigue (least exertion level) and 10 represents maximum fatigue (highest level of exertion). The scale has to be well explained to the subject before the exercise is performed. The advantage of using a subjective measurement is that they are fairly easy to administer as no instrumentation is required and also provides the experimenter with information about the time to interrupt or stop a work test (Borg, 1970). However, the subjective ratings might not reflect only the variables being measured. For example, it is not necessary that the values obtained in an objective measurement comply with the subjective measurement.



The Borg General Scale

0 --	nothing at all
0.5 --	extremely weak (just noticeable)
1 --	very weak
2 --	weak
3 --	moderate
4 --	somewhat strong
5 --	strong
6 --	
7 --	very strong
8 --	
9 --	
10 --	extremely strong (almost maximal)

Figure 1.2. The Borg General Linear Scale

1.1.3. Lower extremity fatigue and gait characteristics

The strength of lower extremity (hip, knee and ankle) plays an important role in stabilizing the whole body during different phases of a gait cycle (Ferber et al., 2002). Apart from normal walking, substantial joint moments (Hip, knee and ankle) are essential for successful recovery from a slip event (Liu et al., 2006). As there are substantial losses due to work-related fall accidents worldwide (Leamon and Murphy, 1995; National safety council, 1995), it is important to understand the factors that contribute to such falls. Localized muscular fatigue of the lower extremity is one such factor that might affect the ability of the lower extremity joints to generate sufficient power to recover from a fall.

The major muscles stabilizing different lower extremity joints are the rectus femoris (quadriceps) and hamstring controlling the knee extension and flexion whereas the gastrocnemius and tibialis anterior controlling the ankle plantar and dorsiflexion. 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 effects of fatigue of the above mentioned muscles have been investigated in many studies (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 group. 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 the knee proprioception (Lattanzio, 1997; Balestra, 1992). Along with the degradation of the proprioception, knee stiffness is also reported to be compromised following a fatigue exertion (Wilson et al., 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 which is a result of fatigue of the stabilizing musculature (Troop, 1988; Lundin, 1993).

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). 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 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). Investigations of older individuals' gait characteristics by Lockhart et al. (2000) and Winter (1990) revealed that the risk of slip-induced falls were 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 extending the knee. 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., 2001). Fatigue induced contractile process and excitation-contraction coupling failure may decrease the hamstring activation leading to higher heel contact velocity.

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 risk of slip initiation. A likely factor influencing the transitional acceleration of the whole body COM may be the ankle plantar flexors' biomechanical and physiological factors – i.e., plantarflexors produce more than half of the positive work during the push-of phase of the gait cycle (Winter, 1983). 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.

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 and 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 - this momentum is 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 risk of falls. Evidence in support of this hypothesis comes from a number of investigations indicating a decline in voluntary muscle strength and rate of muscle force production, and increased 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.

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. Figure 1.3 illustrates a summary of parameters which might be affected by localized muscular fatigue based on the literature review.

As indicated, initial gait characteristic such as slower transitional acceleration of the whole body COM and higher heel contact velocity may affect RCOF due to the increase in horizontal foot force. If localized muscular fatigue affects the initial gait characteristics, the potential for slip-induced falls might 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. One or more of the following sensory inputs may initiate this alerting process: proprioception, vision, and vestibular function. 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 and increase 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 risk of falls.

Though there has been evidence on the effect of gait parameters on the outcome of slips, and also on the effects of fatigue on gait parameters, the relationship between localized muscular fatigue and slip initiation and recovery remains unclear. Fatigue is common during physically demanding work, and the knowledge gained can thus be used to develop practical interventions aimed at improving balance and also reducing the risks of slip-induced falls.

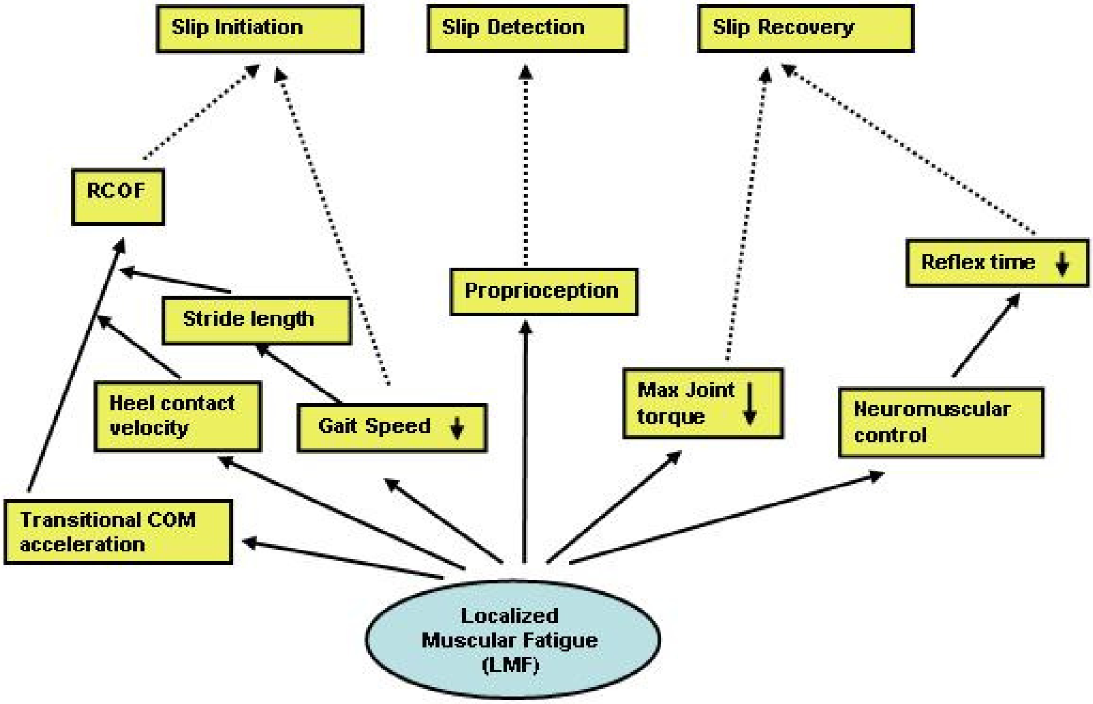


Figure1.3. Effect of localized muscular fatigue on different kinetic, kinematic and neural parameters which affect outcome of slips and falls

1.2. Research Objective and Specific Aims

Occupational fall accidents remain a major cause of considerable number of injuries and fatalities among the American workforce. Also, the exposure to fall hazards has been indicated as a constant aspect of employment (NIOSH, 2000). As such, the accident statistics presented earlier may remain unchanged unless substantial progress is made towards new means of reducing occupational fall occurrences. A majority of these incidents are a result of foot slippage experienced mostly by laborers, and fatigue appears 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.

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 potential risk factors affecting slip initiation and recovery (fatigue), and interventions that promote safe walking.

The overall objective of the current study is to investigate the effects of the localized muscular fatigue of the quadriceps on the slip initiation and slip recovery phases of slips and falls. The improved understanding of the relationship between localized muscle fatigue and slip outcome will enhance our capability to (a) identify localized muscle fatigue as a potential risk factor for slips and falls accidents, and (b) to develop effective intervention (work/rest cycle schedule, exercises) to minimize the cost and rate of injury and death associated with slips and falls. The proposed work addresses several NORA priority areas in context of work-related traumatic falls, like (1) risk assessment methods and (2) Intervention effectiveness research.

Specific Aim #1: To quantify the effects of localized muscle fatigue on slip propensity and fall recovery during slip perturbations. Fatigue (as measure by the reduction in maximum torque produced) will be induced in the knees to assess its effects.

1.2.1. Primary Hypotheses

1. Localized muscular fatigue will adversely affect the gait characteristics; more specifically increase the heel contact velocity and reduce whole body COM velocity and transitional acceleration of the body COM.
2. Localized muscular fatigue will increase the required coefficient of friction thereby affecting the slip propensity.
3. Localized muscular fatigue will increase the slip distances (SDI & SDII) and peak sliding heel velocity (PSHV) during the slip trials in fatigue session, thereby affecting the recovery.
4. Localized muscular fatigue will reduce the peak knee joint moment at the heel contact phase of the gait cycle and during slip resulting in a fall.
5. Localized muscular fatigue will increase the frequency of falls.

Chapter 2

2.0. Methods

Data collection was performed at the Locomotion Research Laboratory, Virginia Tech. The participants were tested for two different sessions, fatigue (F) and no fatigue (NF) within a period of a week. The variables used to quantify the effects of fatigue were heel contact velocity (HCV), transitional acceleration of the whole body centre of mass (TA), required coefficient of friction (RCOF) , velocity of the body COM or walking velocity (WV), slip distances (SDI and SDII), peak sliding heel velocity (PSHV) and peak knee joint moment ($\text{kneemom}_{\text{peak}}$). The experiment was completely randomized for the two different conditions.

2.1. Participants

Sixteen healthy young (18-26 yrs old) adults (10 males & 6 females) participated in the study. Informed consent was approved by the Institutional Review Board (IRB) of Virginia Tech and was signed by all the participants. The participants did not have any musculoskeletal injury or recent injuries (within 6 months) that may adversely impact their ability to perform the fatiguing exercises. Selection criteria for inclusion in the study are presented in more detail under the Human subjects' approval- IRB section in Appendix. The participants' demographic details (age, weight and height) are provided in Table 2.1. All participants reported that they were right side dominant.

Table 2.1 Participant Information

Participant	*Age (yrs)*	*Weight (kg)*	*Height (cm)*
n=16	24.66 ± 3.58	65.86 ± 10.93	174.86 ± 7.67

2.1.1. Sample size estimate

The sample size required for the experiment was estimated from the results of the pilot study. The inter subject variability in the horizontal heel contact velocity during fatigue and no fatigue states was used to calculate the power for the study. Power of the test was determined by focusing on sample sizes large enough to detect differences between fatigue and no fatigue states with high probability. The formula for calculating the power is provided below (Neter et al., 1996)

$$\text{Power} = P\{|t^*| > t(1 - \alpha/2; n\text{-}2 \mid \delta)\}$$

where δ is the noncentrality measure, which is a measure of the distance between the means of A and B (Peak Horizontal heel contact velocity in the fatigue and no fatigue condition):

$$\delta = \frac{|A - B|}{\sigma\sqrt{2/n}}$$

The difference between A and B (i.e. the minimum difference which is important to detect with high probability) was assumed to be 75cm/s, and the standard deviation of the horizontal heel velocity was approximately 50 cm/s. Based on the α value of 0.05, 16 participants should be sufficient to detect the specified differences with risk of Type I error of 0.5 and Type II error of <0.25 (Power > 0.75).

2.2. Data Collection

2.2.1. Apparatus

The apparatus was similar as used in previous slip and fall experiments (Lockhart et al., 2003). 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). Both the force plates were connected to an analog-to-digital converter and amplifier interfaced with LabVIEW data collection software (LabVIEW 6.1, National Instruments™). Kinetic data from force plate were collected at a sampling rate of 1200 Hz. The walkway was covered with baseline vinyl tile (Armstrong). The slippery surface was covered with water and jelly mixture (1:1) to reduce the coefficient of friction (COF) (dynamic COF was 0.12). The experiment layout is shown in Figure 2.1. Uniform experimental shoes were provided to participants to minimize shoe sole differences. 23 markers were placed over the various bony landmarks (Head, Ear, Shoulder, Acromion, Elbow, Wrist, Knuckle, Asis, Knee, Ankle, toe and heel). The marker configuration of this whole body model is illustrated in Figure 2.2. A six-camera ProReflex system (Qualysis) was used to collect three-dimensional position data of the participant while walking. A fall arresting rig was used to prevent participants from hitting the ground once they walked on the slippery surface (Lockhart et al., 2002). Kinematic data (marker data) from the camera were sampled and recorded at 120 Hz.

Biodex Dynamometer (Biodex Medical Systems, Inc., Shirley, NY) was used to induce fatigue and to measure strength by performing isokinetic exercises of the knee. A special attachment was constructed for inducing fatigue in both the knees (Figure 2.3). A visual feedback was provided to the participant while they performed the exertions. Matlab 7.0.1 programming was utilized for all the data analysis and illustrations. SAS 9.1 and JMP 5.1 were utilized for the statistical inferences.

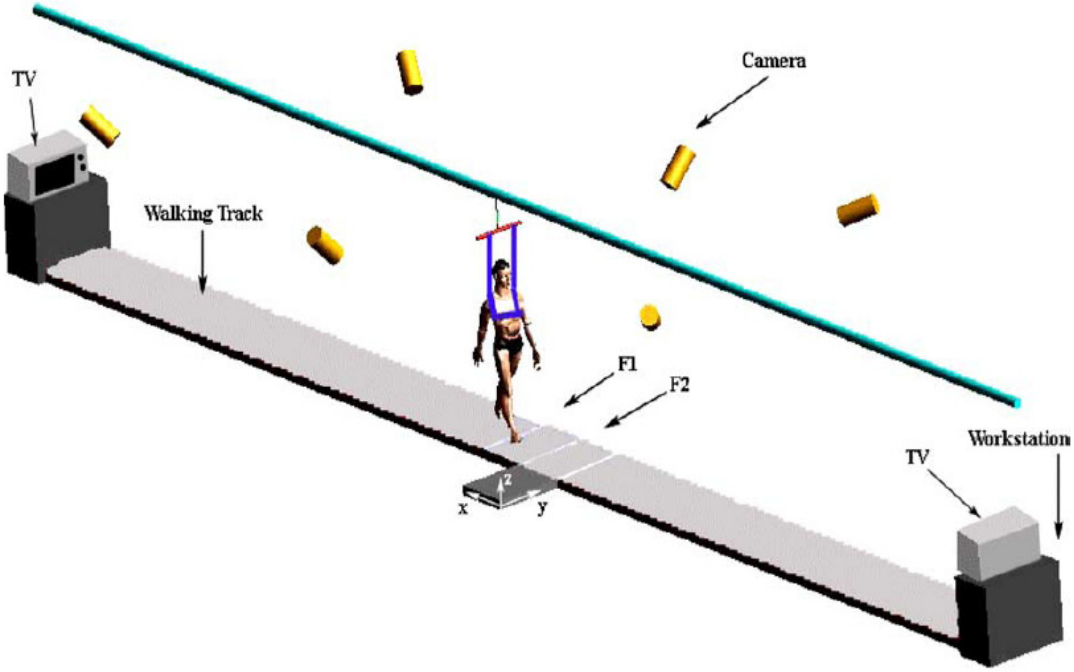


Figure 2.1. Field Layout of the experimental set-up including fall arresting harness, infrared cameras (6), two force plates (F1 & F2), and workstations. X,Y and Z= global references for force and position (Adopted from Kim et al., 2005)

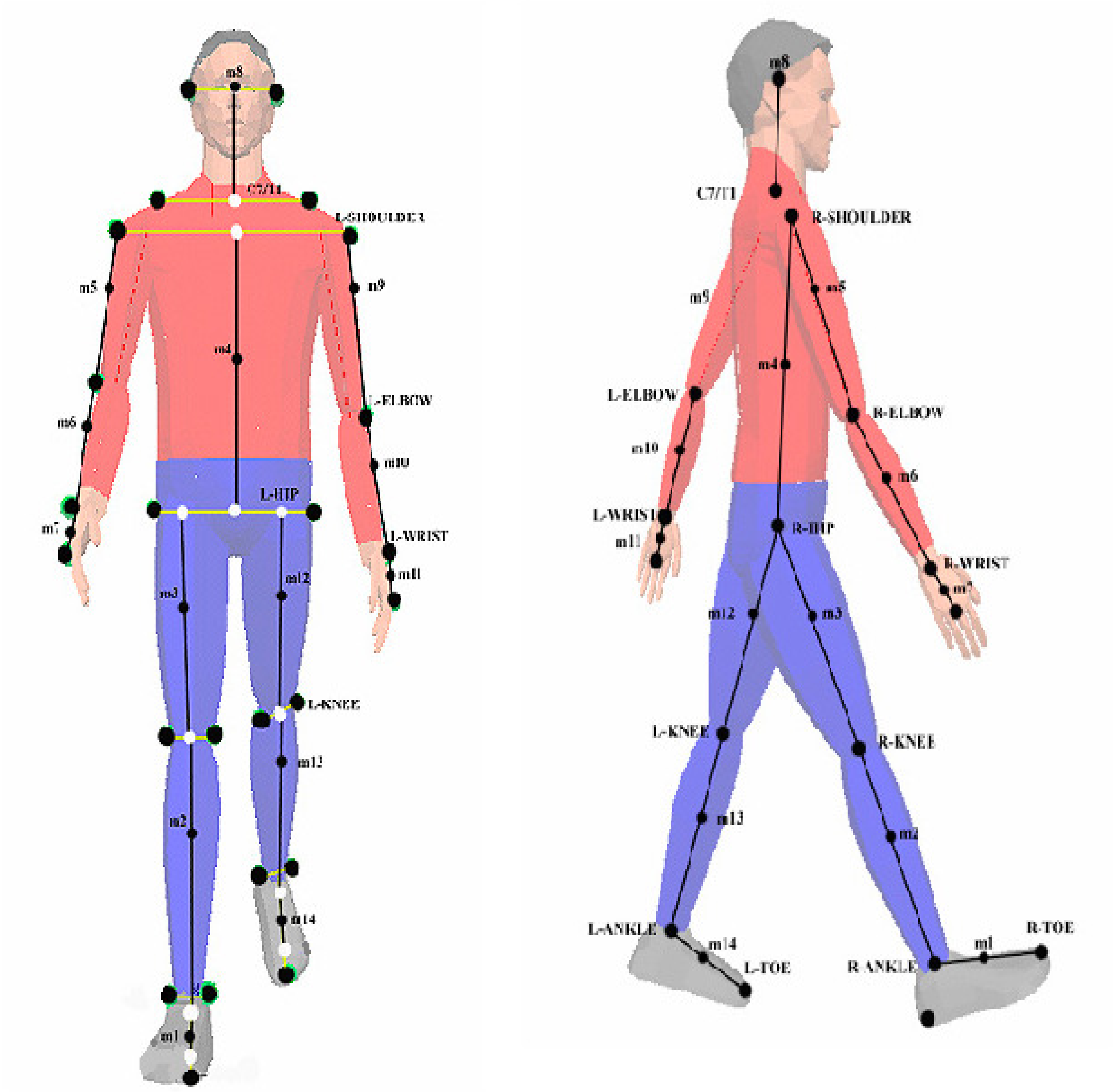


Figure 2.2. Reflective marker placements and internal landmarks (frontal and sagittal view) (Lockhart et al., 2003)

Figure 2.3. Biodex Set up for the isokinetic bilateral fatigue exertions

2.3. Procedure

Participants were involved in two separate sessions within a week. During the no fatigue (NF) session, the fall arresting system and walking instructions were introduced to each participant during the initial familiarization session. Reflective markers were then attached to the anatomical landmarks. During the experiment, participants were instructed to walk normally across a dry vinyl floor surface for 20 minutes. Force plate and the marker data were then collected once the participants felt comfortable and walked with a normal gait. After collecting the normal walking data, participants were asked to perform some simple tasks (filing papers) at both the ends of the linear walkway while they walked. The tasks required them to look at the television screen while they walked and hence helped to remove their attention from the floor. They were also provided with headphones to limit their hearing. Within this 20 minute session, a slippery surface was introduced without participants' awareness and the kinetic data along with the kinematic data were collected simultaneously for the second time now for the slip trial.

During the Fatigue (F) session, participants were first familiarized with the set up and were asked to perform the isokinetic exertions of the knee on the Biodex before the actual experiment. After the markers were attached, the participants walked on the floor surface for about 10 minutes and performed the tasks at both the ends of the linear walkway. Both force plate and marker data were collected in the normal walking condition. After walking, Biodex was utilized to induce fatigue in both their knees.

The fatigue inducement procedures were similar to those recently described by Yaggie and McGregor (2002). Participants were allowed to perform a 5 minute warm up on the Biodex and then their MVE (maximum voluntary exertion) baseline measure was recorded. Bilateral fatigue was induced in both the knees of the participant by having them perform repetitive isokinetic exercises using the extensor muscles of the knee. All exertions were performed at 60^{o}/sec, a value consistent with earlier fatigue protocols (Kay et al., 2000). Participants performed the exertions repeatedly against a resistance set at 70 % of their determined MVE. Visual feedback was provided to the participants for their current and target moment levels. An MVE was performed at regular intervals until the participants reached 60 % of their original MVE and this was considered as the fatigue state. Verbal encouragement was provided to the participants throughout the fatigue protocol for them to do their best. Participants were asked to report a subjective exertion levels (scale 1:10) indicating their perceived fatigue. Immediately after the fatiguing protocol, the participants were asked to walk across the dry vinyl surface and data were collected to represent their fatigue walking. After collecting the fatigue walking data, slippery surface was introduced without paticipants' knowledge within 5-10 min of their walking after the fatiguing protocol. Both kinetic and kinematic data were recorded simultaneously. This was used for the fatigue slip trial analysis. Based on the existing evidence (Nardone et al., 1997; Yaggie and McGregor, 2002), full recovery of the stabilometric measures is expected within about 20 minutes but this holds true for quiet standing after fatigue. To avoid confounding effects due to recovery from imposed muscle fatigue, it is very important to minimize the time between fatiguing exercise and data collection.

To verify the recovery time, another study was conducted. In this pilot study, six healthy young adults performed the same fatiguing protocol as described above and then they were asked to walk on the linear walkway. 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 on the floor again. This was repeated until the participants recovered their original MVE. The study showed different recovery patterns for each individual but none returned to their original MVE before 10 minutes (shown in Figure 2.4). Therefore, this time period was used as a references time window for data collection.

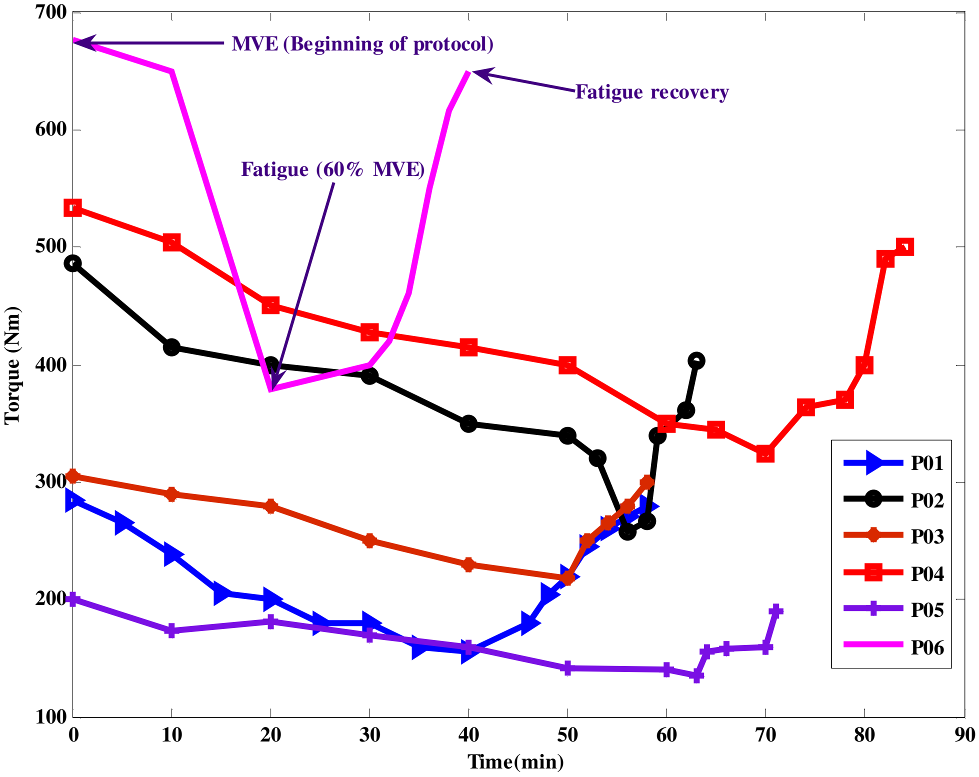


Figure 2.4. Preliminary results of the recovery time from fatigue (60% MVE) compared within six subjects.

2.4. Experiment Variables

2.4.1. Independent Variables

Walking conditions (within subjects) was one of the independent variable. It will have two trials, normal walking (on dry surface) and walking on the slippery surface.

Fatigue status (within subjects) was the second independent variable including the Fatigue and No Fatigue status.

2.4.2. Dependent Variables

The dependent variables are the gait parameters as described below.

1. Heel Contact velocity (HCV):

   It is defined as the instantaneous horizontal heel velocity (HV) at heel contact. It is calculated by numerically differentiating the position data of the heel using the video motion analyses before and after the heel contact:
   $HV = [X_{(i+1)} - X_{(i-1)}]/2\Delta t$.

2. Required coefficient of friction (RCOF):

   It represents the minimum required coefficient of friction between the shoe and floor interface to prevent slipping (Perkins, 1978). This is given as the peak of the ratio between the horizontal and vertical ground reaction force (F_h/F_v) which is calculated from ground reaction forces. This parameter has been linked to risk of slip initiation (Figure 2.5, peak 3 as defined by Perkins, 1978). The peak 1 in the Figure 2.5 is representative of the instant at the heel contact when there is a slight backward motion of the heel. For the purpose of this study, backward slips are analysed and therefore the Peak 3 right after the heel contact is more indicative of RCOF. There is also a peak after Peak 3 which is indicative of RCOF during forward slips.

3. The whole body COM velocity (Also referred as walking velocity (WV)):

   The whole body COM was first calculated by taking a mean of all the centers of mass from the 14 segments as described by Lockhart et al. (2003). This includes left and right feet, left and right shanks, left and right thighs, trunk, left and right hands, left and right upper arms and head.

The formula for calculating the whole body COM velocity is given below:

$$COMvelocity(WV) = [X(i+1) - X(i-1)]/2\Delta t,$$

where X = position of whole body COM

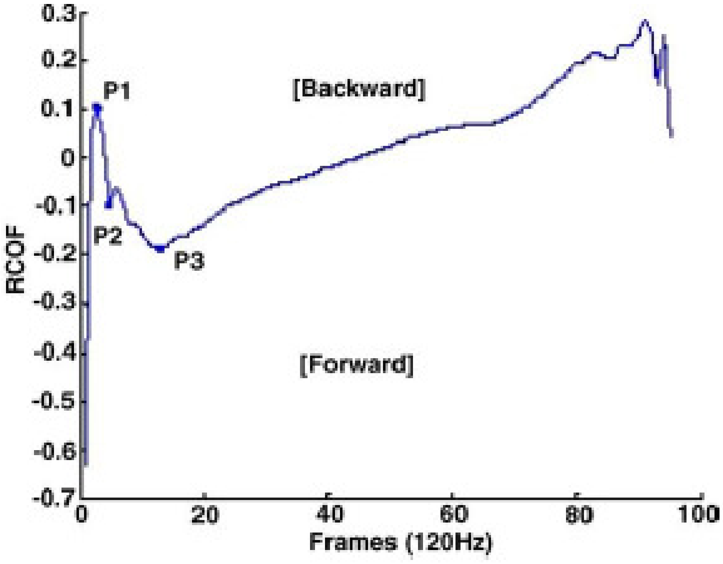


Figure 2.5. Required coefficient of friction (RCOF) peaks during normal gait. Peak 3 is denoted by P3 and occurs shortly after heel contact (adopted from Kim et al., 2005)

4. Transitional acceleration (TA) of the whole body COM

The transitional acceleration of the whole body COM was defined as the change in the horizontal COM velocity between heel contact phase and shortly after heel contact phase of the gait cycle (Lockhart et al., 2003). The change was calculated as the relative horizontal whole body COM velocity differences over time between the whole body COM velocities 50 ms ($\approx$ 7 frames) before and after heel contact (Lockhart et al., 2003). A 3-D link (14) segment model was utilized to calculate position and velocities of the whole body center of mass. It was assumed that the whole body COM in each individual was approximately calculated by the same 14 component link segment system. The transitional acceleration of the whole body COM is used to describe the relationship between the speed of the whole body COM transfer and RCOF differences. It is calculated using the following formula:

$$\dot{v}_{COM} = (v_{COMb} - v_{COMa})/\Delta t.$$

where a and b are position of the COM velocity 50 ms before and after the heel contact.

5. Slip distances

   a. Initial Slip distance (SDI):

      Slip distances are divided into two parts, slip distance I and slip distance II. Initial slip distance or SDI is indicative of severity of slip initiation. Figure 2.6 illustrates a typical slip behavior starting from heel contact phase (Lockhart et al., 2000). In the initial stages, the horizontal heel position (Figure 2.6a) is such that the heel does not slip forward. In approximately 60 ms, the heel starts to slip forward. There is a quick deceleration of the heel (Figure 2.6b) as the horizontal heel velocity decreases. The slip distance is usually defined as the resultant slip distance traveled by the foot after the heel contact phase of the gait cycle.

      The slip distances were obtained using heel co-ordinates between the slip start and slip stop points on the slippery floor surface. The slip start point for the SDI is defined as the point when there was a non-rearward positive acceleration of the heel after heel contact. It is also equivalent to the point where first minimum of the heel contact velocity occurs (Lockhart et al., 2000). The slip stop point for the SDI is defined as the point where peak horizontal heel acceleration occurs after the slip start point (mid slip shown in Figure 2.6). SDI was obtained using the heel coordinates between slip start and slip end point on the floor surface.

b. Slip distance II (SD II):

   Slip distance II is indicative of the behavior of the slip after the slip initiation. Thus, the starting point for the SDII is SDI slip stop point (peak heel acceleration, mid slip shown in Figure (2.6) and the end point is where the slip ends (Lockhart et al., 2002). The slip end is defined as the point where the first maximum of the horizontal heel velocity occurs after the slip start point (Figure 2.6a and 2.6b) (Lockhart et al., 2002). SDII was also calculated from the heel coordinates using the distance between the two points as with SDI.

6. Peak sliding heel velocity (PSHV):

The peak sliding heel velocity is defined as the peak heel velocity after the slip start point. This is indicative of the slip severity after the slip initiation (Figure 2.6). The sliding heel velocity was calculated by averaging the instantaneous sliding heel velocity during the slip-start point to slip-stop point (Khuvasanont, 2002). The horizontal heel velocity was first obtained from the position data. The linear finite difference method used for calculating the velocity utilizes the difference of the foot displacements of the last 1/120 second (Δt) before and after the heel contact divided by the time (2(Δt). It uses the following formula

$$V = \Delta x / \Delta t$$

$$V_i = \left(X_{i-1} - X_{i+1}\right) / 2\Delta t$$

where x is the horizontal distance at frame i and t is 1/120 second per frame.

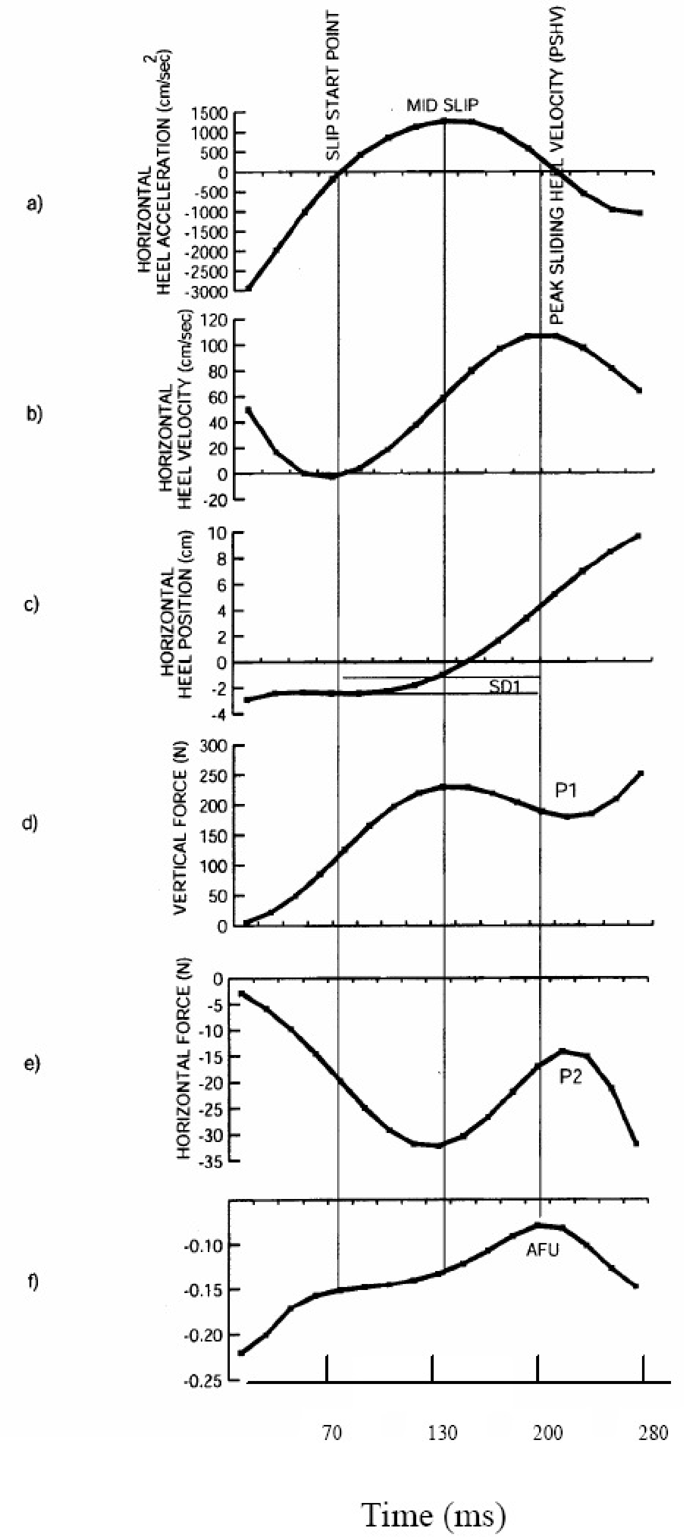


Figure 2.6. Composite view of the gait parameters. (Adapted from Lockhart et al., 2003); these characteristics were found throughout all the participants.

7. Peak joint moment (kneemom$_{peak}$):

An inverse dynamics approach was adopted for the joint moment determination (Liu et al., 2006). Position data were collected in a global coordinate system. Two-dimensional net joint moment of the knee was determined by using Newton's Law of Motion. An example calculation of joint torque is illustrated below (Figure 2.6). The peak value of the knee joint moment (Kneemom$_{peak}$) will be utilized for the analysis along with the pattern. As the knee joint motion is typically in the sagittal plane, all the analysis was be performed in the sagittal plane. The joint moment calculation was performed using the marker data. Figure 2.7 illustrates a typical joint moment profile during normal stance phase of the gait cycle.

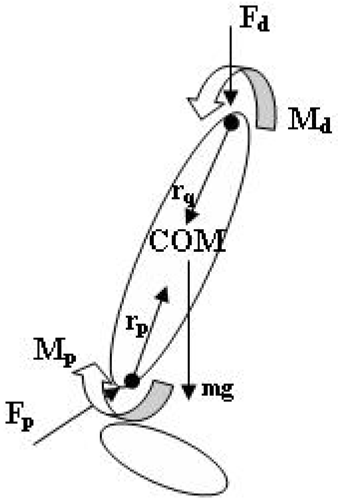


Figure 2.7. Example of joint moment calculation.

In the Figure 2.7, the distal end is denoted as "d" and the proximal end is denoted as "p", segment COM is assumed to lie on a straight line connecting the two endpoints, and if

r_q is the vector from the segment's distal end to its COM and,

r_p is the vector from the segment's proximal end to its COM, then

$$F_p + F_d + mg = ma$$
$$M_p + M_q + r_p \times F_p + r_q \times F_d = I\alpha$$

where I = segment's moment of inertia,

a = linear acceleration of the COM,

α = segment's angular acceleration

All the parameters are treated as vectors.

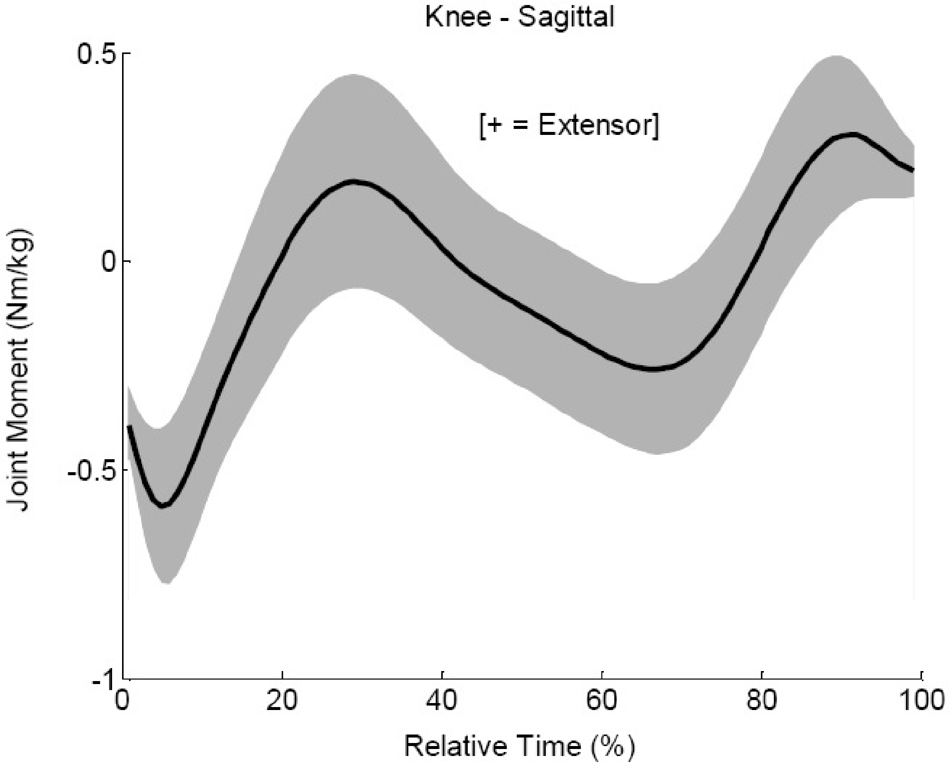


Figure 2.8. Typical knee joint moment profile during the stance phase of the gait cycle (adopted Liu, 2004)

8. Subjective measure on the level of fatigue:

   Subjective assessments for the level of fatigue were obtained using Borg's Scale. The linear rating scale consists of ratings from 0 to 10. The rating 0 being not tired at all and 10 being the highest level of tiredness (want to take a break). The rating scale was explained to the participants well before the task and they were asked to stop doing the exercise at any point they felt uncomfortable. Also, it was made clear to the participants that the use of decimal places was allowed and the value of the ratings did not depend on the value of previous rating.

2.5. Data treatment

The converted coordinate kinematic (marker data) and kinetic (force plate) data were first filtered using a fourth order, zero lag, low-pass butterworth filter. The kinetic data were down sampled to match the marker data. Matlab 7.0.1 was used to perform the processing. There were four sets of data for each participant; two sets of data (normal walking and slip) each for the fatigue and no fatigue session. This was considered a within subject design where the same subjects were used before and after a treatment condition (fatigue and no fatigue). The participants were randomly assigned to these two conditions. Within each of these conditions, the participant experienced normal walking and perturbed walking (slip). Therefore, the data can be treated as pre fatigue and post fatigue. Essentially, we are examining how fatigue might alter the gait characteristics in normal walking and during slip.

Dependent variables were preprocessed by normality check and outlier detection. Box-and-whisker plot and residue plot were applied to assist for any outlier detection.

2.6. Data Analysis

The descriptive and inferential statistics were performed in the JMP and SAS statistical packages. The level of significance was set at 0.05. A one way ANOVA repeated measures analysis was used to test for significant differences between the normal and the fatigue status. A multivariate analysis was performed to examine the correlation between the different dependent measures.

The variables like the RCOF, WV, HCV, TACOM, and $\text{kneemom}_{\text{peak}}$ were analyzed using one way ANOVA within subject design for the normal walking conditions comparing pre and post fatigue sessions. The slip related variables like the slip distances (SDI & SDII), PSHV, and the peak knee joint moment were analyzed for the slip/recovery conditions comparing the pre and post fatigue sessions. The walking velocity and step length was analyzed using one way ANOVA to check if the participants varied their normal gait between the two sessions (NF vs. F) and the two trials (normal vs. slip).

Chapter 3

3.0. Results

The results are categorized into 5 groups examining the gait parameters and friction demand characteristics, slip parameters, peak joint moment during fatigue and no fatigue, ratings of perceived exertion (RPE) and the fall frequency respectively. Also, the relationship between the different dependent variables was summarized using multivariate analysis in the last part of the results section. The gait parameters and friction demand characteristics were obtained from the normal gait trial during fatigue and no fatigue session. The slip parameters were calculated from the slip trials in both the no fatigue and fatigue sessions. The knee joint moment were calculated for both the trials (normal and slip) in both the sessions (No fatigue and Fatigue). The RPE was recorded while the participants performed the knee exertions in the fatigue protocol, and the fall frequency was evaluated using both the slip parameters and the video data.

3.1. Fatigue effects on gait parameters

Table 3.1 provides the summary of the gait parameters and friction demand characteristics (heel contact velocity (HCV), required coefficient of friction (RCOF), walking velocity (WV), and transitional acceleration of the whole body COM (TA) during fatigue and no fatigue states.

The one way ANOVA indicated that the participants walked with a higher HCV in the fatigue state as compared to no fatigue state ($F_{(1,31)}$ =33.86, p=0.0001) (Figure 3.2). It was also observed that the RCOF was higher while walking after fatigue ($F_{(1,31)}$ =9.73 p=0.004). In general, participants walked with a slower speed after fatigue exertions but the difference were not statistically significant ($F_{(1,31)}$ =1.52, p=0.08). Consistent patterns of RCOF were observed during both the conditions in all the participants (Figure 3.1).

The TA in the forward direction was observed to be slower during the fatigue state as compare to the no fatigue state ($F_{(1,31)}$ =3.85, p=0.04). This was evident in all the participants (Figure 3.3).

Table 3.1 Summary of gait parameters (HCV-heel contact velocity, WV- walking velocity, RCOF-required coefficient of friction, TA- whole body COM transitional acceleration)

Variables	*No Fatigue*	*Fatigue*	*ANOVA*
	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/sec2)	199.21(41.27)	159.27(57.16)	*

* $p<0.05$

NS: not significant

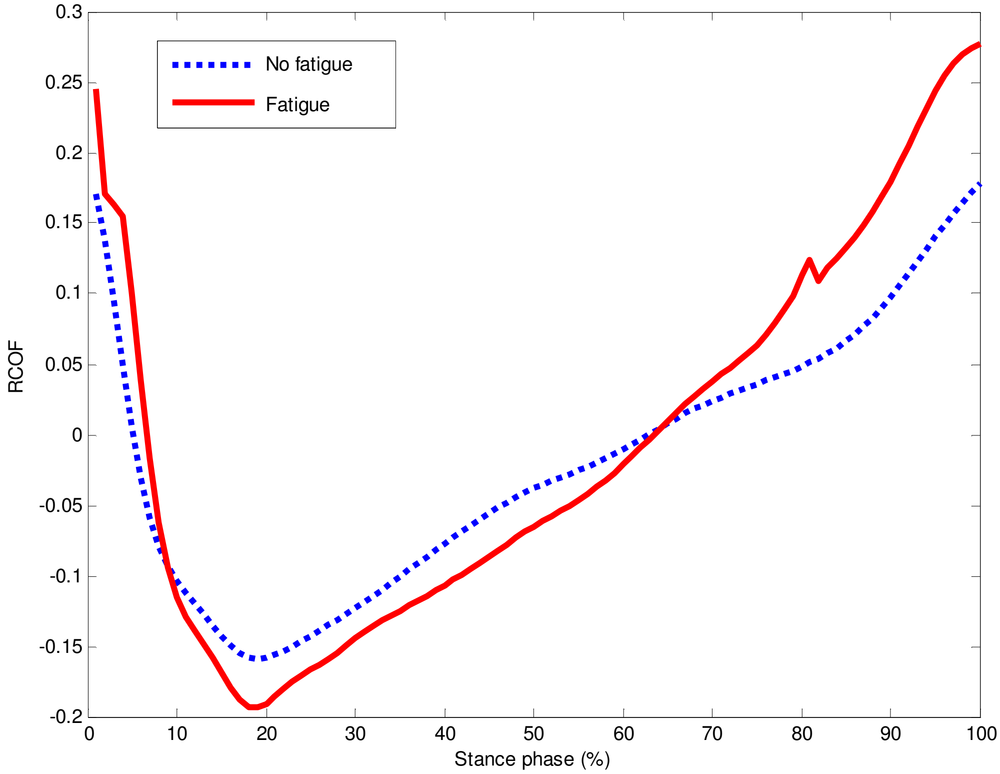


Figure 3.1. Ensemble average of the RCOF of the participants during the two conditions (fatigue and no fatigue- normal walking trials)

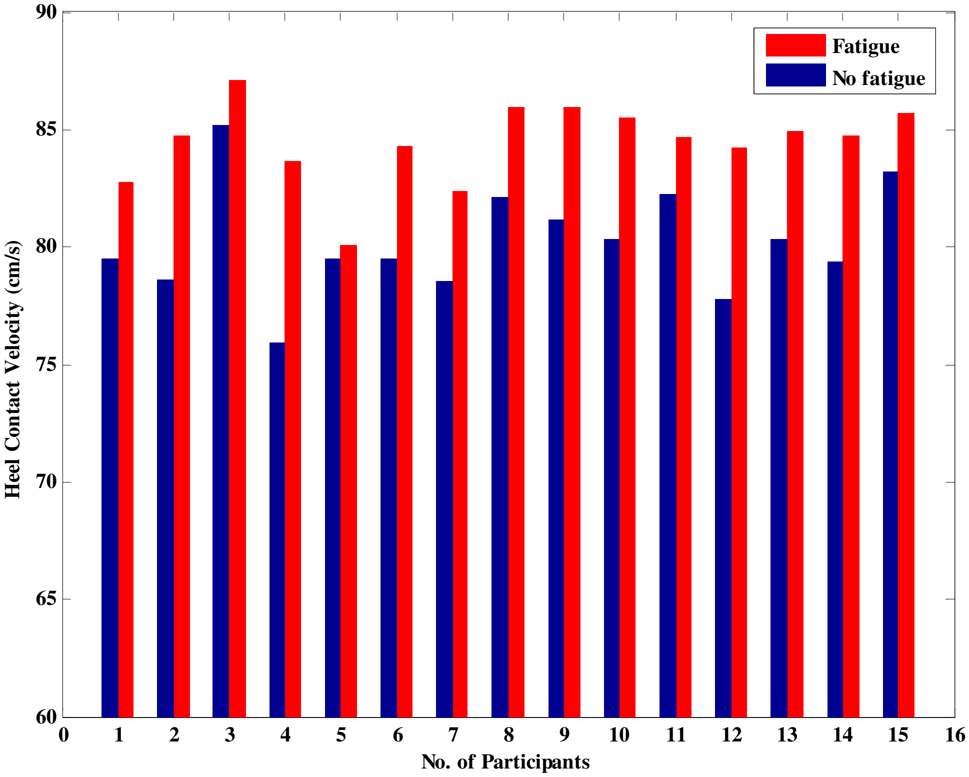


Figure 3.2. Heel contact velocity (HCV) during no fatigue and fatigue session (normal trial).

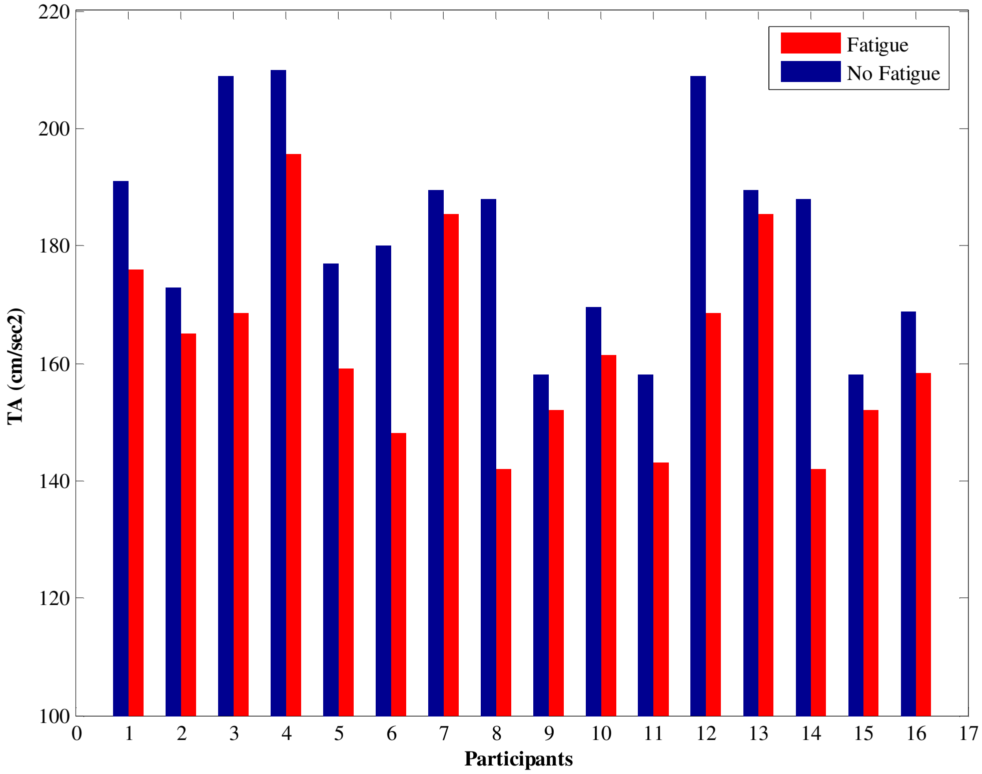


Figure 3.3. Transitional acceleration of the whole body COM (TA) during fatigue and no fatigue session (normal trials).

3.2. Fatigue effects on slip parameters

The slip distances (SDI &SDII) and the peak sliding heel velocity (PSHV) were grouped under the slip parameters as they are indicative of the slip severity (recovery/fall). A summary of the mean and the standard deviation of these parameters in presented in Table 3.2. The one way ANOVA indicated that the participants had longer slip distances both SDI and SDII during fatigue slip trials as compared to no fatigue slip trials. The difference in SDI between the two conditions was not statistically significant (($F_{(1,31)}$ =3.11, p=0.09). SDII was found to be significantly higher during fatigue slip trial as compared to the no fatigue slip trial ($F_{(1,31)}$ =15.19, p=0.008) (Figure 3.4). It was also observed that the PSHV was higher when the participants slipped in the fatigue state, although not statistically significant ($F_{(1,31)}$ =1.08, p=0.12).

Table 3.2 Summary of slip parameters (SDI & SDII -slip distances, PSHV - peak sliding heel velocity)

Variables	*No Fatigue*	*Fatigue*	*ANOVA*
	Mean(S.D)	*Mean(S.D)*	
SDI (cm)	1.8(8.98)	4.1(6.45)	N.S
SDII (cm)	3.7(3.58)	14.3(8.65)	*
PSHV (cm/s)	136.32(69.13)	161.16(53.85)	N.S

* $p<0.05$

N.S-not significant

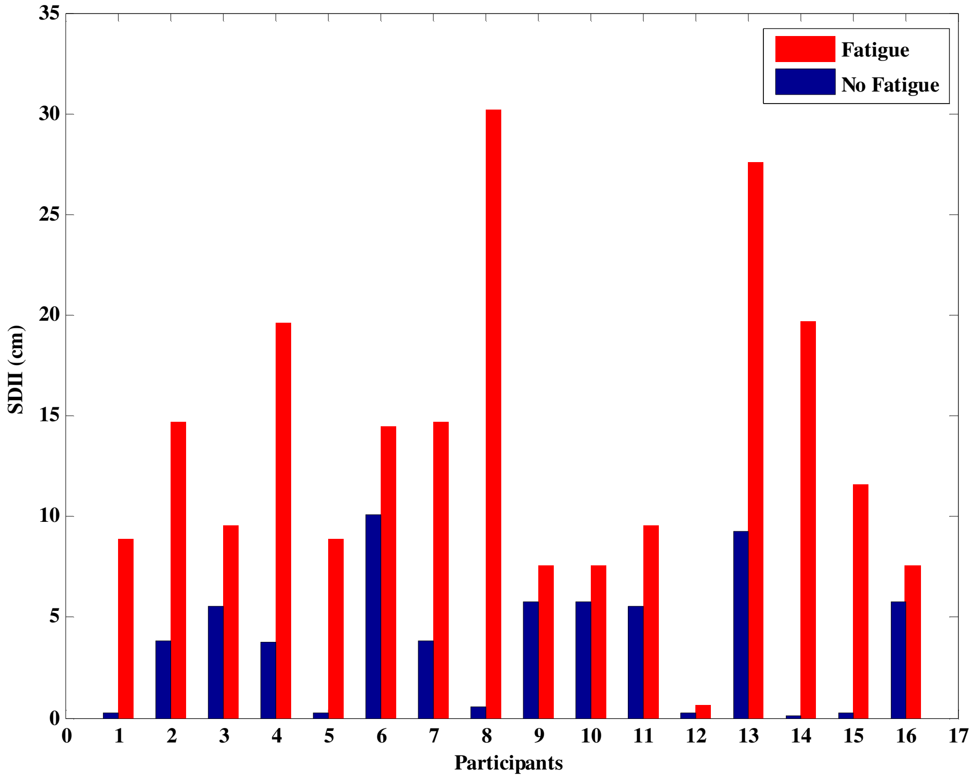


Figure 3.4. Slip distance II during no fatigue and fatigue session (slip trials)

3.3. Effect of fatigue on knee joint moment

The knee joint moments calculated were normalized to the body weight. In the normal walking trials in both no fatigue and fatigue sessions, consistent knee joint moment patterns were observed in all participants (Figure 3.5). At knee joint, sagittal joint moment showed flexor and extensor moment alternatively. Three peaks (P1, P2 and P3) were analyzed for significant differences in both no fatigue and fatigue sessions. All the peaks were analyzed for significant differences. For most of the participants, all the three peaks were present, whereas for some participant only P1 and P2 were present. For simplifying the analysis for Peak 3, only those participants were included who exhibited this pattern (n=12). The peak knee joint moment (P1-extensor) for normal walking trial in the fatigue session was lower than in the no fatigue session, but the difference was found to be statistically insignificant ($F_{(1,31)}$ =3.06, p=0.09). Similarly, for the Peak 2, the differences were found to be statistically insignificant ($F_{(1,31)}$ =3.28, p=0.08). The Peak 3, which was again extensor dominated, was found to be significantly different in both the states ($F_{(1,23)}$ =7.02, p=0.01). The P3 (extensor moment) was lower in the fatigue session as compared to the no fatigue session. The Table 3.3 summarizes the mean and standard deviation of the peak joint moment during normal walking trials in both fatigue and no fatigue session.

The knee joint moment profile in Figure 3.6 is an ensemble average of all the participants who had a reactive recovery. It is the average of the knee moment at each point starting at heel contact to toe off after the slip. As seen in the Figure 3.4, the joint moment was predominantly extensor moment. The ANOVA performed on the recovery trials from slips in both the conditions (fatigue and no fatigue) 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). The Table 3.4 summarizes the mean and standard deviation of the peak knee moment ($\text{kneemom}_{\text{peak}}$) during the two sessions for normal and slip trials.

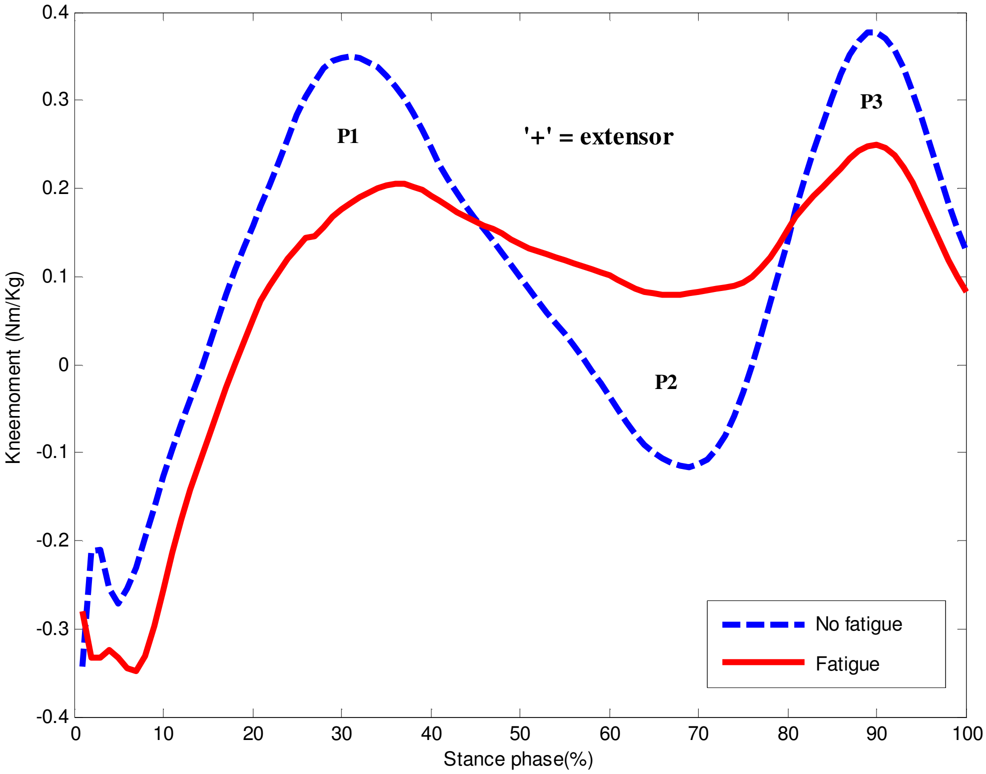


Figure 3.5. Average stance phase joint moment profile at knee sagittal plane during no fatigue and fatigue walking trials .

Table 3.3 Mean and standard deviation of the peak knee moment (kneemompeak) P1, P2 and P3, during the no fatigue and fatigue sessions (normal trial)

Session	**Kneemom$_{peak}$ (Nm/Kg)**		
	Peak 1 Mean(SD)	**Peak 2** Mean(SD)	**Peak 3** Mean(SD)
No Fatigue (Normal)	0.45(0.19)	0.15(0.29)	0.45(0.26)
Fatigue(Normal)	0.23(0.22)	0.04(0.07)	0.23(0.20)

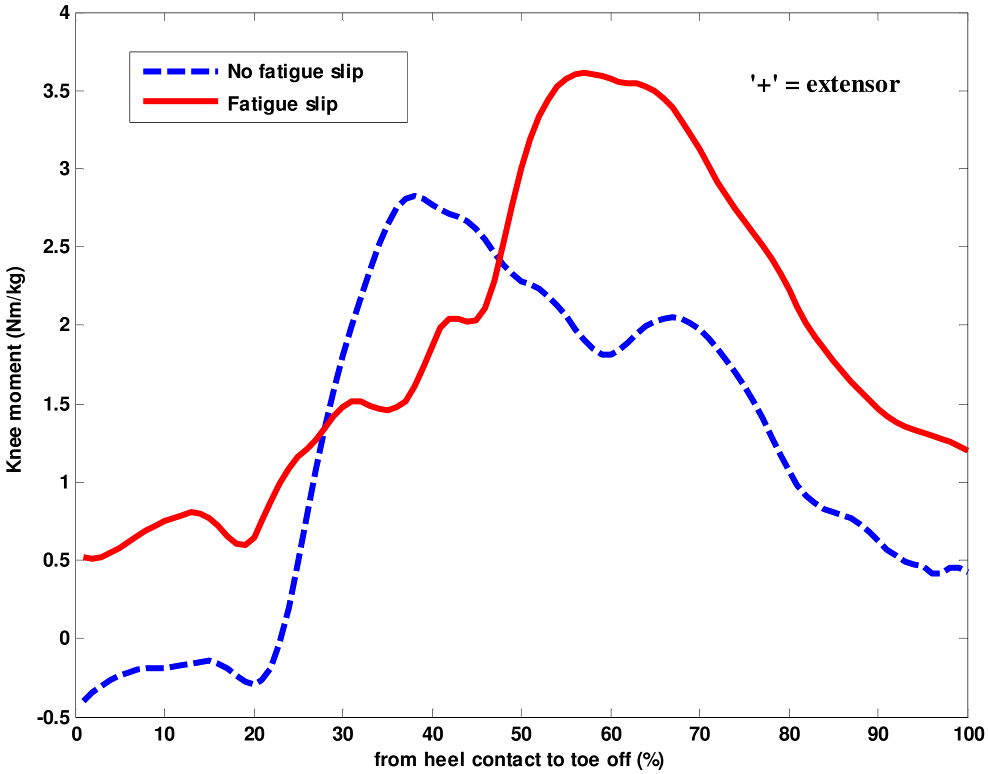


Figure 3.6. Average knee joint moment profile during reactive recovery in no fatigue and fatigue session (slip trial).

Table 3.4 Mean and standard deviation of the peak knee moment (kneemompeak) during the no fatigue and fatigue sessions for the slip trials

Condition	**Kneemom$_{peak}$ (Nm/kg)** Mean (SD)
No fatigue (Slip)	3.44(0.97)
Fatigue(Slip)	5.25(2.83)

3.4. Walking velocity and Step length

The walking velocity and the step length during the normal walking trials in both the sessions (NF and F) were analyzed using one way repeated measure ANOVA to check if the participants changed their natural gait between both the sessions. The normal walking data during the fatigue session refers to the data collected before inducing fatigue. There was no significant effect of the sessions on the walking velocity ($F_{(1,31)}$ =3.52, p=0.083) and step length ($F_{(1,31)}$ =0.1053, p=0.747). The Table 3.5 summarizes the mean and standard deviation of walking velocity and step length in normal walking trials during NF and F trials.

Table 3.5 Mean and standard deviation of walking velocity and step length during normal walking in 2 sessions, fatigue (F) and no fatigue (NF)

	Session 1 (NF) Mean(SD)	Session 2 (F) Mean(SD)
Walking velocity (cm/s)	127.83(44.23)	131.75(87.89)
Step Length (cm)	68.53(6.89)	70.24(7.38)

Also, in order to confirm that the participants did not change their gait within the same session between the two trials (normal and slippery), one way repeated ANOVA was performed on the step length and the walking velocity. The results are presented in Table 3.6. The normal and slippery data presented in the F session refers to the data collected after fatigue protocol. There was no significant effect of the trials in NF session on the walking velocity ($F_{(1,31)}$ =0.406, p=0.53) and step length ($F_{(1,31)}$ =0.35, p=0.88). Also, there was no significant effect of the trials in F session on the walking velocity ($F_{(1,31)}$ =0.249, p=0.62) and step length ($F_{(1,31)}$ =1.159, p=0.29).

Table 3.6 Mean and SD of walking velocity and step length during trial 1 (normal) and trial 2 (slip) in each of the two sessions (NF- no fatigue and F-fatigue)

Variables	NF Session		F Session	
	Trial 1 (Normal) Mean(SD)	Trial 2 (Slippery) Mean(SD)	Trial 1 (Normal) Mean(SD)	Trial 2 (Slippery) Mean(SD)
Walking velocity (cm/s)	127.83(44.23)	119.47(19.07)	120.26(20.59)	124.49(20.88)
Step Length (cm)	68.53(65.89)	79.86(38.63)	68.82(6.48)	75.45(18.54)

3.5. Frequency of falls

There were four falls in the fatigue session and one fall in the no fatigue session. Various parameters were utilized 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., 2002). Also, the videos for each of the participants were analyzed to detect a fall.

3.6. Subjective measures on the level of fatigue

RPE was obtained from each participant for the level of exertion at an interval of 5 minutes and the average values at each interval were plotted for all the participants. The changes in RPE are shown as a function of time (Figure 3.7). It was observed that most of the participants reached their level of maximum exertion by 40-45 min.

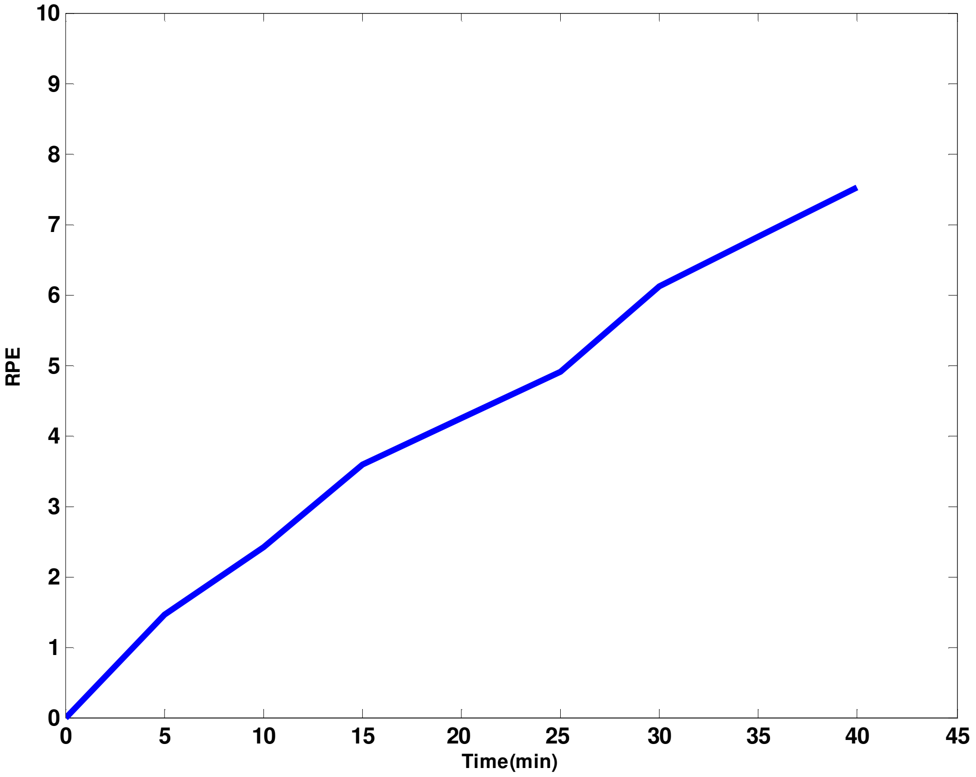


Figure 3.7. Average changes in ratings of perceived exertion (RPE). The RPE indicated perception of discomfort in the lower extremity (knee).

3.7. Summary of correlation analysis

A multivariate analysis was performed to examine the correlation between different dependent variables. The Table 3.7 summarizes the pair wise correlation between various dependent variables. The analysis indicated that HCV and RCOF were highly correlated (R^2=0.39, p=0.026). Also, there was a negative correlation between HCV and TA (R^2= -0.37, p=0.03). In terms of slip parameters, the analysis revealed a positive correlation between HCV and slip distance II (SDII) (R^2=0.4, p=0.04). There was also a strong positive correlation between SDII and $\text{Kneemom}_{\text{peak}}$(R^2=0.53, p=0.007). The Figures 3.8, 3.9, 3.10, and 3.11 represent the bivariate fit of each pair of variables having strong correlation.

Table 3.7 Pairwise correlation of different dependent variables in the study (multivariate analysis), (WV-walking velocity, RCOF- required coefficient of friction, TA- transitional acceleration of the whole body COM, RCOF- required coefficient of friction, SDI- slip distance 1, SDII- slip distance II, PSHV- peak sliding heel velocity)

Variable	***By Variable***	***Correlation***	***Sig. Prob.***
WV (cm/sec)	HCV(cm/sec)	-0.1355	0.4597
RCOF	HCV(cm/sec)	0.3936	**0.0258**
RCOF	WV (cm/sec)	-0.2843	0.1148
TA	HCV(cm/sec)	-0.3799	**0.0320**
TA	WV (cm/sec)	0.2274	0.2107
TA	RCOF	-0.0842	0.6468
SDI	HCV(cm/sec)	0.2398	0.2591
SDI	WV (cm/sec)	0.2775	0.1893
SDI	RCOF	0.2402	0.2582
SDI	TA	0.1680	0.4327
SDII	HCV(cm/sec)	0.4253	**0.0383**
SDII	WV (cm/sec)	-0.0111	0.9591
SDII	RCOF	0.3609	0.0832
SDII	TA	-0.0162	0.9400
SDII	SDI	0.3151	0.1337
PSHV	HCV(cm/sec)	0.0384	0.8587
PSHV	WV (cm/sec)	-0.2240	0.2928
PSHV	RCOF	0.0132	0.9511
PSHV	TA	-0.2085	0.3282
PSHV	SDI	0.0017	0.9938
PSHV	SDII	0.1465	0.4947
$\text{Kneemom}_{\text{peak}}$	HCV(cm/sec)	0.1826	0.3931
$\text{Kneemom}_{\text{peak}}$	WV (cm/sec)	0.0859	0.6898
$\text{Kneemom}_{\text{peak}}$	RCOF	0.0792	0.7131
$\text{Kneemom}_{\text{peak}}$	TA	0.2320	0.2754
$\text{Kneemom}_{\text{peak}}$	SDI	0.2718	0.1989
$\text{Kneemom}_{\text{peak}}$	SDII	0.5343	**0.0072**
$\text{Kneemom}_{\text{peak}}$	PSHV	-0.1327	0.5366

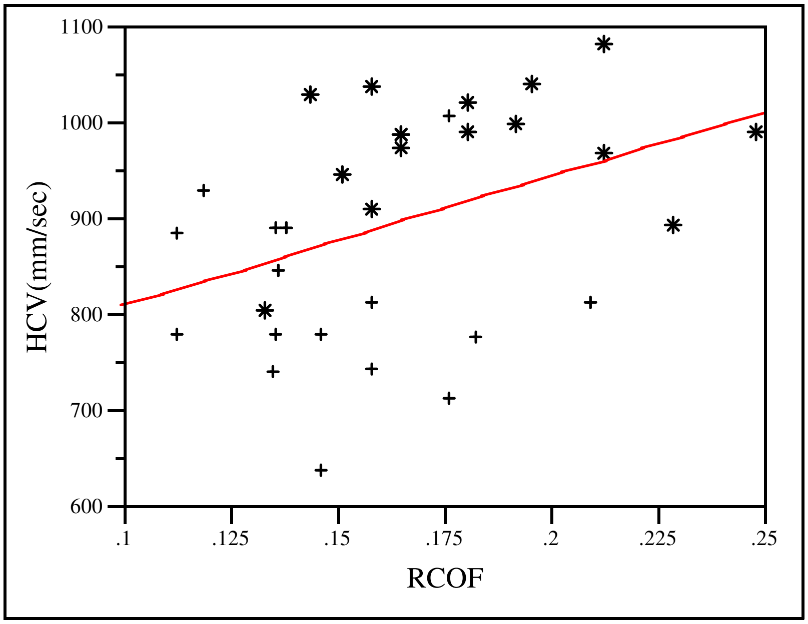


Figure 3.8. Bivariate fit of heel contact velocity, HCV (mm/s) by required coefficient of friction, RCOF ('+'=No fatigue, '*'= Fatigue)

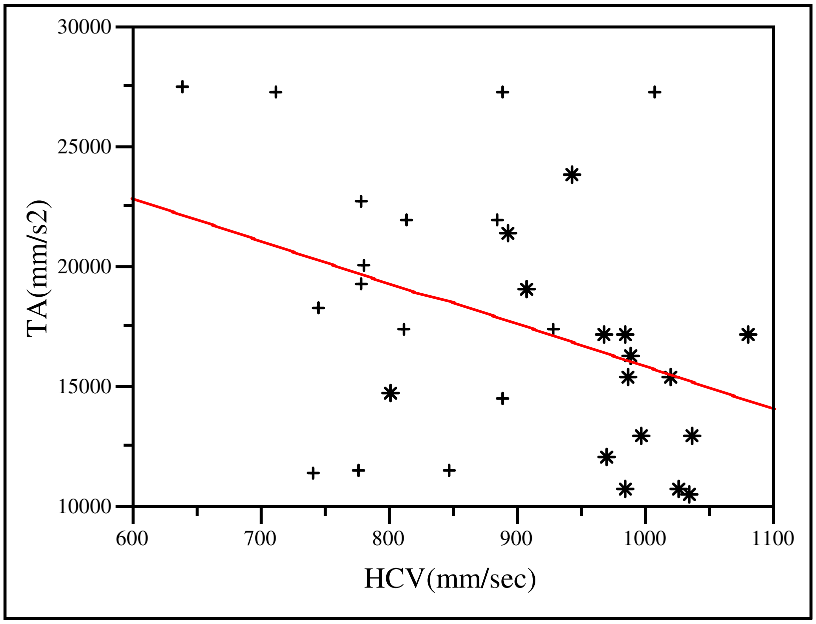


Figure 3.9. Bivariate fit of transitional acceleration TA (mm/s2) by heel contact velocity, HCV(P< 0.03, R2= 0.14) ('+'=No fatigue, '*'= Fatigue)

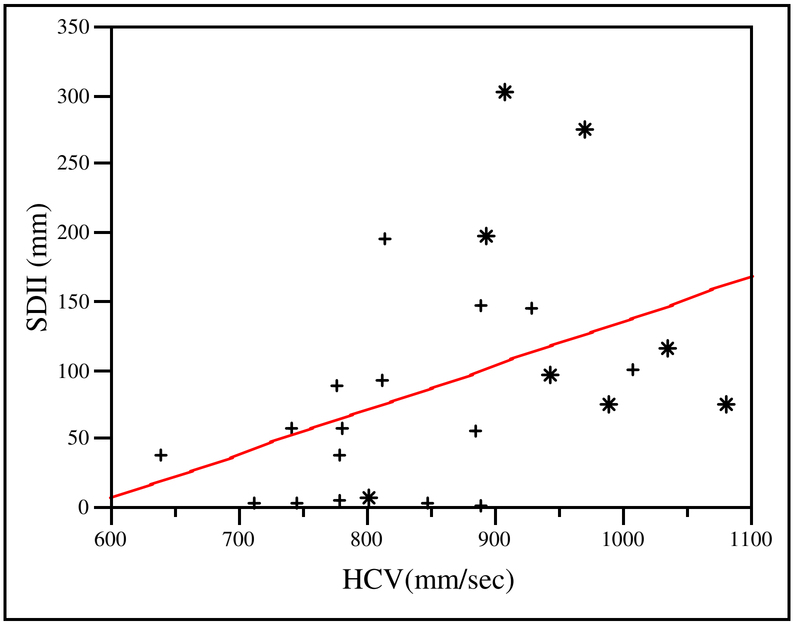


Figure 3.10. Bivariate fit of slip distance (SDII) by heel contact velocity, HCV (mm/s) (‘+’=No fatigue, ‘*’= Fatigue)

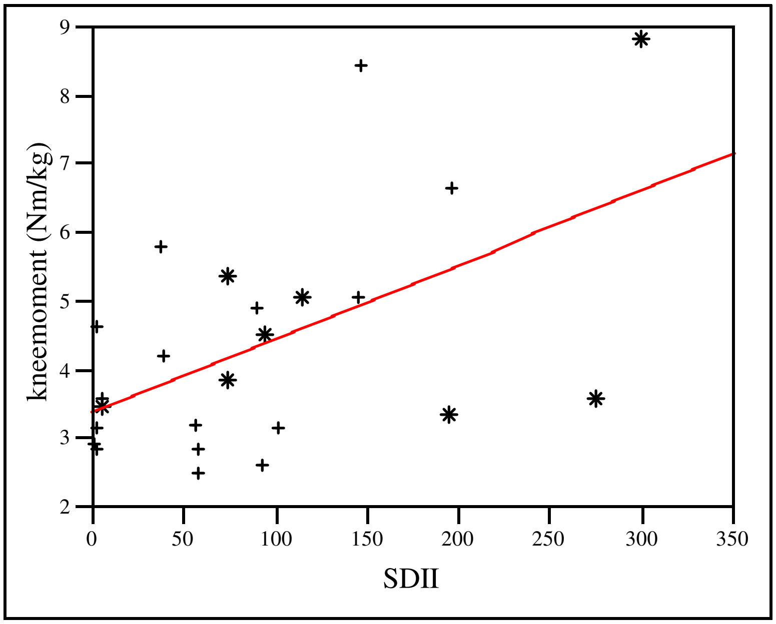


Figure 3.11. Bivariate fit of knee moment peak by slip distance II (SDII) (mm) (‘+’=No fatigue, ‘*’= Fatigue)

Chapter 4

4.0. Discussion

The purpose of this study was to examine the effects of localized muscular fatigue of the knee on the different gait parameters and friction demand characteristics along with the slip parameters. The results have shown that fatigue of the knee adversely affects the gait and is also responsible for increasing the slip propensity. Each of the hypotheses and the corresponding findings will be discussed in detail in this chapter.

4.1. Hypothesis 1: "Localized muscular fatigue will adversely affect the gait characteristics; more specifically increase the heel contact velocity and reduce whole body COM velocity and transitional acceleration of the body COM."

The heel contact velocity (HCV) is simply defined as the heel velocity during the instant of heel contact. It was calculated using the heel marker position data. This variable was expected to reveal the gait characteristics affecting slip initiation.

The HCV is considered important in terms of gait parameters as it can drastically change the friction demands while walking. It has been studied that HCV affects the required coefficient of friction (RCOF) by altering the ratio of horizontal to vertical foot forces (Lockhart et al., 2003). 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 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 were higher due to the higher HCV. It was found in the results of the current study that the localized muscle fatigue of the knee substantially increased the HCV of the individuals. Additionally, the bivariate analysis suggested a strong positive correlation between HCV and RCOF, indicating that the RCOF increased with an increase in the HCV. The results are consistent with the study by Saggini et al. (1998) who examined the effects of localized muscle fatigue on the lower extremity and concluded that fatigue increased the gait cycle time and also increased the horizontal heel velocity. A likely factor influencing the higher heel contact velocity may be a decrease in the hamstring activation rate. The 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 heel velocity. An increase in the heel contact velocity was considered to increase the likelihood of slip-induced falls (Karst et al., 1999; Mills and Barrett, 2001). Although implicated, further study is required to clearly understand the increase in the HCV due to fatigue.

The second parameter in the hypothesis was the transitional acceleration of the whole body COM (TA). This term was defined by Lockhart et al. (2003) as the relative horizontal COM differences before and after heel contact. It was calculated to describe the relationship between the speed of the whole body COM transfer and RCOF differences.

During an external perturbation like a backward slip, the speed of the forward momentum of the body is essential and an inability in producing this momentum might 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 when an individual walked after fatiguing the knee, the TA in the sagittal plane was reduced. Syed and Davis, (2000) observed that lower extremity fatigue during walking changed the loading rate and increased ground reaction forces thereby reducing the forward momentum of the whole body COM. Yoshino et al. (2004) also confirmed that the back acceleration (in all the three directions, anterior-posterior, medio-lateral and vertical) reduced due to fatigue induced by prolonged walking. This is in agreement with the current finding. However, it was suggested by Mizrahi et al. (2000) and Voloshin et al. (1999) that fatigued musculature is less able to attenuate heel strike-initiated shock waves, which could be observed as increase in the amplitude of the acceleration measured at the shank and sacrum. The differences in our study might be due to various reasons. One of them being, that in Mizrahi et al. (2000) study, the participants ran on a treadmill at a predetermined speed, whereas in our study they walked over ground at their comfortable pace. Secondly, the discrepancies might be due to the difference in the calculation methods. In the current study, the TA was defined as acceleration before and after heel contact to look at the transitional acceleration. Lockhart et al. (2003) indicated that reduced push-off force of the stance leg further reduced TA and increased RCOF and risk of slip initiation. In other words, a reduction in the TA due to localized muscle fatigue is likely to increase the friction demand at the shoe floor interface of the contacting foot. However, the bivariate analysis between the TA and RCOF indicated no strong correlation. This might be due to independent control of these two variables as the participants walked at their self selected pace. Similar results were observed in a study by Kim et al. 2005 which indicated no correlation between RCOF and TA in older individual's gait. The study concluded that TA was not a direct predictor of RCOF for the older individuals.

The question is if TA is not correlated to RCOF, can it still be considered to affect the slip severity? Although no significant correlation was present between TA and RCOF, the bivariate analysis revealed a strong negative correlation between TA and HCV. This indicates that the reduction in the TA was correlated to the increase in the HCV after fatigue. As discussed earlier, an increase in HCV was related to slip severity. In theory, a reduction in TA should affect the slip recovery due to the lack in producing sufficient forward momentum to be able to recover. Further analysis is required to examine the effects of TA on slip severity after fatigue.

The whole body COM velocity (WV) was calculated using the body COM position. It was hypothesized that the walking velocity would reduce after the fatiguing protocol. Though, the participants walked at a slower speed after fatigue, the decrease in their waking velocity during fatigue were not statistically proven. This can again be contributed to the fact that each participant walked at their self selected speed. A study by Moyer et al. (2006) concluded that the slip severity was higher in individuals who walked with a higher walking velocity. This was in contradiction with the present study, as the walking velocity was lower after fatigue and more falls were reported during fatigue state. This indicates that a number of parameters are responsible for the change in the gait characteristics influencing slip severity.

A study by Yoshino et al. (2004) evaluated the effects of fatigue induced due to prolonged walking on gait parameters. It was concluded that by the end of the experiment, the walking velocity substantially reduced and the gait cycle was longer for all the participants. The current finding supports this study but the discrepancies might be due to differences in the methods for inducing fatigue. The fatigue in the current study was just limited to the localized muscle fatigue of the knee whereas in the other study the participants walked on the treadmill for 90 minutes. So, it can be concluded that fatigue does affect the walking velocity which is again related to any alterations in the other gait parameters.

In summary, the results partially support the hypothesis. They are indicative of slip initiation and are related to the friction demand characteristics of an individual. Localized muscular fatigue of the knee resulted in a higher heel contact velocity and slower transitional acceleration of the whole body COM which are considered as risk factors for slip initiation.

4.2. Hypothesis 2: "Localized muscular fatigue will increase the required coefficient of friction thereby affecting the slip propensity."

The required coefficient of friction (RCOF) or friction demand represents the minimum coefficient of friction that must be available at the shoe-floor interface to prevent slip initiation (Kim et al., 2005). It is characterized by the ratio between horizontal force and vertical force.

In terms of slip-induced falls, friction demand characteristics between the shoe floor surface and the floor has been implicated as an important predictor variable related to severity of falls. In the current study, it was observed that after the fatiguing protocol, participants walked with a higher required coefficient of friction. It has been discussed in the literature review section that higher required coefficient of friction is related to increase in the slip propensity. The reason for a higher RCOF after fatigue can be related to changes in the HCV and TA after fatigue as observed in the previous hypothesis results. 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 will alter the RCOF. Furthermore, Lockhart et al. (2003) indicated that a reduction in the TA is likely to increase the friction demand at the shoe floor interface of the contacting foot, though there was no significant correlation found between TA and RCOF in the current study. Increased initial friction demand (i.e., RCOF) would lead to a higher likelihood of slips associated with low coefficient of friction floor surfaces.

In a bivariate analysis to assess the gait parameter's effect on the RCOF, it was found that HCV was a predictor variable for the RCOF in the older adults and both heel contact velocity and transitional acceleration of COM were predictor variables for RCOF in younger adults (Kim et al., 2005). It is interesting to know that fatigue effects on the gait parameters in the current study are somewhat similar to the effects of aging specifically on RCOF, HCV and WV.

In summary, localized muscle fatigue of the knee resulted in the increase of RCOF and thus was suggested to contribute in the increase of slip propensity during fatigue state.

4.3. Hypothesis 3: "Localized muscular fatigue will increase the slip distances (SDI & SDII) and peak sliding heel velocity (PSHV) during the slip trials in fatigue session, thereby affecting the recovery."

The slip distances (SDI & SDII) and the peak sliding heel velocity (PSHV) are more indicative of fall/recovery as compared to the other gait parameters. The SDI reveals the slip severity at the slip initiation and the SDII explains the slip behavior after the slip initiation. It was observed in the study that the slip distances were significantly longer in the fatigue state as compared to the no fatigue state. The SDI difference between the two states was not statistically different. The SDII however, was significantly longer in the fatigue state and can be contributed to the falls in some of the participants. It is generally accepted that a fall will occur during a slip, if the slipping foot exceeds 10 cm and if the peak velocity of the slipping foot exceeds 50 cm/s (Perkins, 1978). However, it was observed in the current study that participants recovered from a fall even after a slip longer than 10 cm and PSHV higher than 100cm/s in the fatigue state. This result is in agreement with the study by Brady et al. (2000) which concluded that participants recovered with heel velocities exceeding 50cm/s. All the participants who fell in the fatigue state in the current study had an average SDII of 25 cm and all those who recovered had an average SDII of 15cm. Also, as expected the PSHV was higher for participants who fell.

The PSHV was higher during the fatigue slips even though not statistically significant. This probably was due to the large standard deviation in this measure, which indicates subject dependent strategy during a slip and fall. However, the frequency of falls was higher in the fatigue state and high PSHV were observed in most of the participants. The reason for the slip distance II to be longer in case of fatigue might be due to the change in other gait parameters. There was a strong positive correlation between SDII and HCV indicating that increase in HCV might have some effects on the SDII. There was also a strong correlation between the SDII and the peak knee moment generation. It was seen previously that on average, TA was faster in the no fatigue state. This suggests that the slip distances were shorter in no fatigue state as individuals reacted and moved faster to a slip as compared to in the fatigue state. This could also be related to reaction time and as we know that fatigue affects proprioception, there might be a delay in the initiation of the corrective mechanism which resulted in a longer slip distance. Further analysis is required to confirm the relationship between the SDII and the delay in reaction time.

Lockhart et al. (2002) reported that the older adults slipped longer than their younger counterparts and this was due to the difference in their gait characteristics. The study indicated that they had slower walking velocity and higher heel contact velocity and higher RCOF. The changes in the gait parameters here in the current study during the fatigue state are similar to the referenced study and thus it might be concluded that by altering the gait parameters, fatigue can influence the SD and the PSHV. It was also concluded by Brady et al. (2000) that the slip displacement and not velocity, was a determinant of whether slip resulted in a fall. This was also evident in the current study where an increase in the SDII was indicative of more falls in the fatigue state.

In summary, the results partially support the hypotheses. It was evident that changes in the gait parameters due to fatigue resulted in longer slip distances SDII in participants and this can be directly related to the slip severity in terms of a fall or recovery.

4.4. Hypothesis 4: "Localized muscular fatigue will reduce the peak knee joint moment at the heel contact phase of the gait cycle and during slip resulting in a fall."

Characteristic stance phase joint moment profiles were observed in both the normal and fatigue conditions. The average joint moment profiles in normal gait are in accordance with previous literature (Cham et al., 2001; Redfern and DiPasquale, 1997; Eng et al., 1995). The knee joint moment profile during the fatigue state had both reduced extensor and flexor moment. These findings are contradictory to some studies (Bruggemann, 1994; Christina, 2001). The discrepancies might be due to different activities (running, rapid stop) and different fatigue patterns (general lower extremity fatigue, dorsiflexor fatigue). Different fatigue patterns may elicit different biomechanical effects and due to this, compensation strategies might be different. The peak joint moment during the normal walking conditions was higher than that of the fatigue state, though the difference was not statistically significant. The literature provides enough evidence that fatigue reduces the force generating capacity of the lower extremity muscles which in turn reduces the joint torque (Robinovich et al., 2002; Wolfson et al., 1995; Svantesson, 1998). The hypothesis was based on these studies and it was expected that the peak knee moment would reduce after the fatigue state. One of the reasons for the insignificant difference is the huge variability in the knee moment data which might be due to the difference in fatigue level of each participant. In contrary to normal gait, joint moment magnitudes in the slip recovery during both no fatigue and fatigue states were highly variable. Such large variability can be mainly contributed to the individual specific nature of slip/fall accidents which was apparent in previous studies also (Cham et al., 2001).

The current study observed that the peak knee moment during the reactive recovery phase was significantly higher in the fatigue slip-recovery as compared to the no fatigue slip-recovery. The pattern indicated a dominant extensor moment while recovering from a slip in both the sessions (no fatigue slip and fatigue slip). This is contradictory to the results indicated by some studies (Cham and Redfern, 2001). In their study, a considerable increase (relative to normal gait) in flexion moment was observed at the knee between 25% and 45% into stance. Such discrepancies might be for two reasons. Firstly, the time interval adopted for the analysis is different for different studies. In the current study, the targeting interval was from heel contact to toe off, assuming it covered the whole balance recovery. Secondly, in the previous study the subjects were aware of the slippery surface whereas in the current study, unexpected slips were induced without the participant's awareness. However the current results were in agreement with the study by Ferber et al. 2002, which concluded a higher knee extensor moment while recovering from a perturbation.

The peak joint moment was higher during the fatigue slip and this can be explained based on the study by Liu et al. (2006) which indicated higher moment generation requirement while recovering from a slip. It was evident earlier in the discussion that after fatigue, participants had higher SDII, which implied that they slipped longer (> threshold SD =10cm). Also, it has been suggested that higher SDII can directly affect the outcome of slips/falls (slip severity). It can be deduced from this argument that higher joint moment generation would be required to recover from a slip which is more severe (>10cm). Also, the peak knee joint moment was strongly correlated to the SDII in the bivariate analysis, again indicating that higher knee joint moment during a slip might also be related to slip severity. For the participants who fell during the fatigue slip session (n=4), their peak knee joint moment was below 5.25Nm/kg which was the average peak knee joint moment of the participants who recovered.

Also the recovery in the participants in the fatigue slip session can be contributed to the fact that the population in the experiment consists of a mix of participants in terms of strength and this was evident during the fatiguing protocol. While some of the participants fatigued in 30 min, others required 45 min to reach the same level. There is a possibility that people who recovered from a slip were at a higher strength level. It was interesting to look at the knee moment profile, which indicated a rapid rise in moment in the no fatigue slip-recovery as compared to the fatigue slip-recovery. However, the slip occurred within 70-120 ms of the heel contact and therefore the interval chosen for analysis, can have effects on the final result as seen in previous studies. Further analysis is required to examine factors such as joint power, joint moment reaction time, and moment generation at other joints (ankle, hip) to completely explain the knee joint behavior during the slip and fall/recovery.

In summary the results partially support the hypothesis. The major findings of the study indicate a higher peak joint moment generation while recovering from a fatigue slip as compared to recovery from a no fatigue slip. In addition, in normal walking trials, the differences between peak joint moment between no fatigue and fatigue sessions were not significant.

4.5. Hypothesis 5: "Localized muscular fatigue will increase the frequency of falls."

Four of the participants fell in the fatigue condition and there was only one fall in the no fatigue condition. None of the other individuals fell in either of the conditions. It can be concluded that although other individuals also had longer slip distances, not everyone was prone to the fall accidents. This can be due to several reasons, one of them being the fitness of the individuals and also prior knowledge of the slippery condition. As already described, some of the participants were very strong and that was evident during the fatigue protocol, as they generated substantially greater joint moment. In conclusion, the frequency of falling was higher in the fatigue state as compared to the no fatigue state.

Chapter 5

Common Sense is that which judges the things given to it by other senses.
..Leonardo da Vinci

5.0. Conclusion

The purpose of this investigation was to determine the effects of lower extremity fatigue (knee) on the different gait parameters and to examine how fatigue relates to slip severity. The major findings of the study indicated that localized muscular fatigue of the knee alters the important gait and slip parameters which are responsible for recovery from a fall. The study was conducted in two different sessions (no fatigue and fatigue). In both sessions, the participant was tested for walking on a normal surface and walking on an unexpected slippery surface. The different dependent variables were the RCOF, WV(cm/s), TA(cm/s^2), HCV(cm/s), SDI & SDII (cm), PSHV(cm/s) and $\text{kneemom}_{\text{peak}}$ (Nm/kg). It was concluded that fatigue significantly increased the RCOF, HCV, SDII, and $\text{kneemom}_{\text{peak}}$(slip) whereas reduced the WV, TA, and $\text{kneemom}_{\text{peak}}$(normal). In addition, the fall frequency was higher during fatigue slip (n=4) as compared to the no fatigue state (n=1).

Increases in the RCOF, HCV, SDII and $\text{kneemom}_{\text{peak}}$(slip) due to localized muscular fatigue can be considered as potential risk factors for slip severity (fall/recovery). The Figure 5.1 illustrates the relationship between localized muscular fatigue and the different dependent variables. In addition, it represents the correlation between the different dependent variables and how they can be related to the slip initiation and recovery phases. In the multivariate analysis, HCV was the strongest predictor of RCOF and the SDII, so it can be concluded that HCV is particularly important during slip initiation and is affected adversely by localized muscular fatigue of the knee. HCV was also strongly correlated to RCOF and therefore an increase in both these parameters due to fatigue can be contributed to affect the slip initiation process. The increase in SDII and $\text{kneemom}_{\text{peak}}$(slip) due to fatigue can be contributed to affect the slip recovery process. It was hypothesized earlier in the introduction that localized muscular fatigue is a potential risk factor for slip-induced fall accidents. The results partially support the hypothesis. Although localized muscular fatigue did affect the different gait parameters responsible for slip initiation and recovery indicating higher slip severity, it was observed that during an actual slip event more than 50% of the participants (n=11) recovered from the slip after the fatigue protocol. Simultaneously, it was also observed that these individuals required much higher joint moment generation for recovering from the fall in the fatigue state. This indicates a potential risk of injury to the fatigued muscles due to such high joint moment generation. Also, it was observed that more number of participants fell after fatigue protocol (n=4) as compared to the no fatigue protocol (n=1). Further biomechanical analysis is required to clearly predict localized muscular fatigue as the potential risk factor for slip and fall accidents.

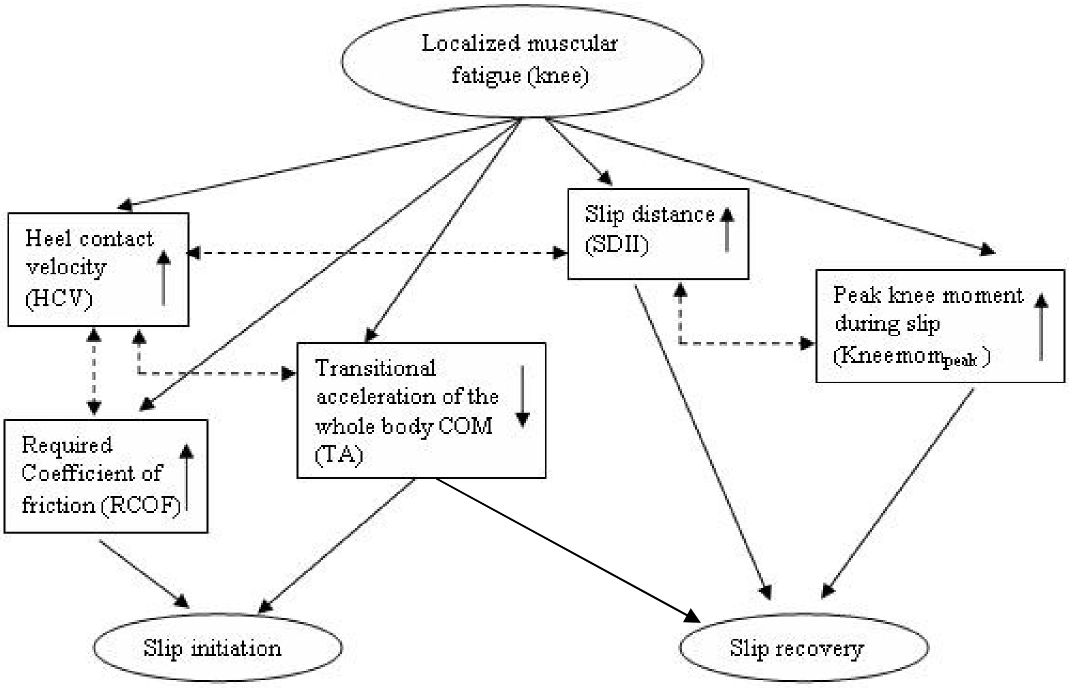


Figure 5.1. Summary of the findings of the study indicating the relationship between localized muscular fatigue and different dependent variables and, the correlation within different dependent variables. ('solid line arrows indicate the direct relationship observed and the dotted line arrows indicate the correlation between the variables')

5.1. Limitations

5.1.1. Possible Contribution of the fatigue level difference

The primary objective of this study was to evaluate the effects of fatigue and hence it was very important to carefully administer this treatment condition to all the participants. The fatigue protocol (isokinetic exertions of the knee) was consistent for all participants (till they reached 60 % of their original MVE). In addition, RPE measurements indicated the subjective level of fatigue. There were two cases when participants had not reached their fatigue level but they gave higher rating of their RPE. This is one of the limitations of the study as these participants had not reached their fatigue level. Also, each participant reached their fatigue level at different time period. Although effort was made to verbally encourage the participants to do their best, some were highly motivated and some were not. This cannot be completely quantified. These limitations can affect the results due to the difference in the fatigue level in each individual. To minimize any effects of the fatigue recovery, the post fatigue trial was conducted within the time frame suggested by the pilot study (< 10 min).

Another limitation in regard to the fatigue protocol is the use of bilateral fatigue. Previous studies on the difference between the strength in unilateral and bilateral exertions have produced equivocal results. Some of such studies suggest that joint strength is reduced in case of bilateral exertion as compared to unilateral exertion (Vandervoort et al., 1984; Khodiguian et al., 2003). However, a study by Jakobi et al. (1998) concluded that the force production is not altered during bilateral contractions. The reason why bilateral fatigue was considered in the current study was to simulate a real condition where both the limbs get fatigued (gait analysis) and not just one.

5.1.2. Prior knowledge of existence of slippery surface and learning effects

The current study utilized a slippery surface to evoke unexpected slips. The primary objective was to simulate a real life situation where slips are experienced unexpectedly and then evaluate the bodily reactions from the participants. For the protection of the participants, they were informed of the existence of a potential slippery surface. This could have led to some gait adaptation (Cham et al., 2002). To minimize the gait adaptations, several precautions were taken. Firstly, the participants were unaware of the location of the slippery surface. Secondly they were asked to do some tasks at both the ends of the walkway, which required them to constantly look at the TV monitor at each ends and also they were made to wear sound proof head phones. Each participant was also allowed a warm up period to get comfortable walking with the harness.

The two different sessions (both fatigue and no fatigue) were performed on two different days within a span of a week. In order to reduce any learning effects, even in the fatigue session, the participants were first asked to walk normally on the pathway and perform tasks before making them go through the fatigue protocol. Also, these sessions were randomly assigned to the participants, as a result of which some participants had the fatigue session first followed by the no fatigue session and vice versa. As the participants experienced slip again in the second session, there might be some learning effect as they had already experienced it in the earlier session. To minimize this effect, the participants were encouraged to walk normally and were given different tasks to do at both the ends of the walkway. Also, the step length and walking velocity of the participants did not differ significantly between both the days, which indicated that the participants did not significantly adapt their gait.

5.1.3. Possible threats to external validity

The population chosen for the study comprised of young healthy college students. Since the motivation for the study was to evaluate the fatigue in workers, the current study had two limitations. Firstly, to be able to refer back the current findings in the students to the workers population and secondly the task simulated for inducing fatigue in the current study was not a typical real world task for workers. The protocol however, was designed to fatigue the knee in a way that it represents the fatigue due to prolonged walking. Workers continuously engage themselves in several different activities throughout the day requiring different muscle/joint movements at different times. Therefore it is difficult to assess their biomechanical requirements in a controlled lab setting.

5.1.4. Possible instrumentation error

The marker data and the force plate data were utilized for all the analysis. The motion data from markers have issues of missing markers (hidden) at some instants during the data collection and this might affect the data. The marker data however was checked initially before the analysis, to make sure all the markers were present. In an event where some major markers were missing, the data were not processed further. Also, during the slip trials, the FP data were not very reliable after the slip start due to individual reactive corrections. As long as the slipping foot was in contact with the FP, the data were reliable. There were instances when participants lifted their slipping foot off the FP and then landed their foot again. To minimize this flaw, the initial data after the slip start was analyzed between the two different sessions.

5.2. Potential application of the research

It was proposed in the introduction of this thesis that the improved understanding of the relationship between localized muscle fatigue and slip outcome will enhance our capability to (a) identify localized muscle fatigue as a potential risk factor for slips and falls accidents, and (b) to develop effective intervention (work/rest cycle schedule, exercises) in order to minimize the cost and rate of injury and death associated with slips and falls. The proposed work addresses several NORA priority areas in the context of work-related traumatic falls, like (1) risk assessment methods and (2) intervention effectiveness research. The study led to a conclusion that localized muscular fatigue of the knee severely affected some important gait parameters and hence can be related to slip severity based on these parameters.

One of the applications of this research is to use the data as preliminary information on the effects of fatigue on some specific gait parameters and fall outcomes. Some conclusions can be drawn from the study regarding which parameter was most affected by fatigue and further in depth analysis of those parameters can be undertaken to confirm the results. Also, the results from the fatigue recovery can be used as a timeframe in which the future experiments should be conducted in order to accurately assess the effects of fatigue (specifically for the knee).

5.3. Recommendations and Future Studies

The recommendations are based on overcoming some of the limitations described earlier. The population consisted of healthy young individuals (10 males and 6 females) in the age group of 18-25 yrs. However, the distribution between the genders was not equal. This makes it difficult to evaluate any gender differences on the effects of fatigue on gait parameters and slip outcome. Equal number of males and females will help to identify gender effects, if any. Participants in the current study walked at their comfortable pace and hence it was difficult to explain the behavior of certain variables which might change with cadence and therefore were subject specific. A controlled cadence rate for all the individuals would help in removing that effect. However, a more controlled experiment would increase the limitations due to the threat to external validity.

Further analysis of joint moment is required to effectively assess the effect of fatigue on joint moment (e.g. joint moment reaction time, joint power etc.). In terms of joint moment, the current study only evaluated the 2D joint moment as the knee motion is predominantly in the sagittal plane, but a 3D joint moment analysis might reveal some results confirming the effects of fatigue. Also, the current study did not evaluate the ankle and hip joint moment primarily assuming that knee joint fatigue might not affect the other joint motions. It would be interesting to evaluate whether the fatigue at one joint alters the function at adjacent joints while recovering from a slip or even during normal gait cycle.

The future studies can be designed to evaluate the effects of both ankle and knee fatigue on the outcome of slips and falls, as ankle joint is also considered important during the gait cycle. In order to effectively evaluate fatigue effects in a particular job, the experiments can be designed simulating real world tasks (e.g. lifting, bending, twisting, walking on inclined surfaces etc.) and then induce unexpected slips to observe the response. Also, future studies can evaluate the effect of fatigue on aging and examine if fatigue is also a potential risk factor for the elderly individuals who are more prone to falling. It would be interesting to evaluate how fatigue influences gait adaptation when a participant is aware of a slippery surface. There might be some differences in voluntarily controlling the different gait parameters (joint motion, walking velocity etc.) when there is a known slippery surface vs. unexpected slips. In the field of intervention research, experiments can be conducted evaluating the effects of rest cycles between the fatiguing activities to observe the changes in the gait parameters.

REFERENCES

Allman, B. L., Rice, C. L. (2002). "Neuromuscular fatigue and aging: central and peripheral factors." Muscle & Nerve **25**: 785-796.

Armstrong, R. B., Taylor, C.R (1993). "Glycogen loss in rat muscles during locomotion on different inclines." J. Exp. Biol. **176**: 135-144.

Astrand, P. O., Rodhal, K. (1986). Textbook of work physiology. New York, McGraw-Hill.

Baker, A., Kostov, KG., Miller, RG., Weiner MW. (1993). "Slow force recovery after long- duration exercise: metabolic and activation factors in muscle fatigue." J Appl Physiol **74**: 2294-2300.

Balestra, C., Duchateau, J., Hainut, K. (1992). "Effects of fatigue in the stretch reflex in a human muscle." Electroencephalogr Clin Neurophysiol **85**: 46-52.

Begg, R. K., White, R., Major, R. E. (1989). "Instrumentation used in clinical gait studies: a review." Journal of Medical Engineering and Technology **13**: 290-295.

Bentley, T. A., Haslam, R. A. (2001). "Identification of risk factors and countermeasures for slip, trip and fall accidents during delivery of mail." Applied Ergonomics **32**: 127-134.

Bigland-Ritchie., W., J. (1986). "Changes in muscle contractile properties and neural control during human muscular fatigue." Muscle & Nerve **7**: 669-699.

Bonder, B. R., Wagner, M.B. (1994). Functional performance in older adults. Philadelphia, F.A. Davis Company.

Borg, G. (1985). An Introduction to Borg's RPE Scale. Ithaca, NY, Movement Publications.

Brady, A. R., Pavol, J. M., Owings, M. T., Drabiner, D.M. (2000). "Foot displacement but not velocity predicts the outcome of a slip-induced in young subjects while walking." Journal of Biomechanics **33**: 803-808.

Bruggermann, G. P., Arndt, A. (1994). Fatigue an dlower extremity function. Proceedings from the Eight Biennial Conference, Canadian Society of Biomechanics.

Cady, E., Jones, DA.,Lynn, J.,Newham, DJ. (1989). "Changes in force and intracellular metabolites during fatigue of human skeletal muscle." J Physiol **418**: 311-325.

Caldwell, L. S. (1974). "A proposed standard procedure for static muscle strength testing." American Industrial Hygiene Association Journal **35**: 201-206.

Chaffin, D. B., Wolstad, J.C., & Trujillo, A. (1992). "Floor/shoe slip resistance measurement." American Industrial Hygiene Association Journal **53**: 283-289.

Cham, R., Redfern, M.S. (2001). "Lower extremity corrective reactions to slip events." Journal of Biomechanics **34**: 1439-1445.

Cham, R., Redfern, M.S (2002). "Changes in gait while anticipating slippery floors." Gait & Posture **15**: 159-171.

Chang, W. R., Kim, I.J., Manning, D. P., Bunterngchit, Y. (2001). "The role of surface roughness in the measurement of slipperiness." Ergonomics **44**: 1200-1216.

Christina, K. A., WHite, S.C., Gilchrist, L.A. (2001). "Effect of localized muscular fatigue on vertical ground reaction forces and ankle joint motion during running." Hum. Move. Sci. **20**(3): 257-276.

Clancy, E. A., Cairns, K.D., Patrick, O., Melvin, B.S., Kerrigan, D.C (2004). "Effect of treadmill walking speed on lateral gastrocnemius muscle firing." Am J Phys Med Rehabil **83**(7): 507-514.

Cohen, H. H., Lin, L. (1991). "A retrospective case-control study of ladder fall accidents." Journal of safety research **22**: 21-30.

Corbeil, P., Blouin, S.J., Begin, F., Nougier, V., Teasdale, N. (2003). "Perturbation of the postural control system induced by muscular fatigue." Gait & Posture **18**: 92-100.

Courtney, T. K., Webster, B.S. (2001). "Antecedent factors and disabling occupational morbidity - insights from the new BLS data." AIHAJ **62**: 622-632.

Courtney, T. K., Sorock, G. S., Manning, D. P.,Collins, J. W., Holbein-Jenny, M. A. (2001). "Occupational slips, trips, and fall-related injuries- can the contribution of slipperiness be isolated?" Ergonomics **44**: 1118-1137.

Davis, J. M. (1996). "Carbohydrates, branched-chain amino acids and endurance: the central fatigue hypothesis." Sports Science Exchange **9**(2): 61.

De Luca, C. J. (1984). "Myoelectric manifestations of localized muscular fatigue in humans." Critical Reviews in Biomedical Engineering **11**: 251-279.

Do, M. C., Brenier, Y., Brenguier,P. (1982). "A biomechanical study of balance recovery during a fall forward." Journal of Biomechanics **15**: 933-939.

Dohm, A. (2000). "Gauging the labor force effects of retiring baby boomers." Mon Labor Rev **July**: 17-25.

Eng, J. J., Winter, D.A. (1995). "kinetic analysis of the lower limbs during walking:what information can be gained from a three-dimensional model?" Journal of Biomechanics **28**: 753-758.

Ferber, R., Osternig, R.L., Woolacott, H. M., Wasielewiski, J.M., Lee, H.J (2002). "Relative balance adjustments to unexpected perturbations during human walking." Gait & Posture **16**: 238-248.

Fitts, R. H. (1996). "Muscle fatigue: the cellular aspects." The American Journal of Sports and Medicine **24**(6): 9-13.

Forestier N, T. N., Nougier V (2002). "Alteration in the position sense at the ankle induced by muscular fatigue in humans." Med Sci Sports Exer **34**: 117-122.

Fullerton, H. (1999). "Labor force projections to 2008: steady growth and changing composition." Mon Labor Rev **November**: 19-32.

Fullerton, H. N., Toosi, M. (2001). "Labor force projections to 2010: steady growth and changing composition." Mon Labor Rev **November**: 21-38.

Gauchard, G., Chau, N., Mur, J.M., Perrin, P. (2001). "Falls and working individuals: role of extrinsic and intrinsic factors." Ergonomics **44**: 1330-1339.

Gefen, A. (2002). "Biomechanical analysis of fatigue- related foot injury mechanisms in athletes and recruits during intensive marching." Med. Biol. Eng. Comput. **40**: 302-310.

Gerdle, B., Langstrom, M. (1987). "Repeated isokinetic plantar flexions at different angular velocities." Acta Physiol Scand **130**: 495-500.

Gronqvist, R., Roine, J., Jarvinen, E., Korhonen, E. (1989). "An apparatus and a method for determining the slip resistance of shoes and floors by simulation of human foot motions." Ergonomics **32**(8): 979-995.

Hakkinen, K., Komi, P.V (1986). "Effects of fatigue and recovery on electromyographic and isometric force- and relaxation- time characteristics of human skeletal muscle." Eur J Appl Physiol Occup Physiol **55**(6): 588-596.

Harper, F. C., Warlow, W. J., and Clarke, B. L. (1967). "The forces applied to the floor by the foot in walking. II. Walking on a slope." National Building Studies Research Paper Ministry of Technology (London: HMSO): 1-8.

Hoon-Yong Yoon, T. E. L. (2006). "Nonfatal occupational injuries associated with slips and falls in united states." International journal of industrial ergonomics **36**: 83-92.

Horak, F. B., Diener, H. C., Nashner, L. M. (1989). "Influence of central set on human postural responses." J Neurophysiol **62**: 841-853.

Hortobagyi, T., Lambert, N.L., Kroll, W.P (1991). "Voluntary and reflex responses to fatigue with stretch- shortening exercise." Can J Sports Sci **6**: 142-150.

Hsiao, H. S., P. (2001). "Preventing falls from roofs: a critical review." Ergonomics **44**: 537-561.

Karst, G. M., Hageman, P.A., Jones, T.V., Bunner, S.H., (1999). "Reliability of foot trajectory measures within and between testing sessions." J Gerontol **5A**: M343-347.

Kay.D., S. C. G., A., Mitchell, M.J.,Lambert, M.I., and Noakes, T.D. (2000). "Different neuromuscular recruitment patterns during eccentric, concentric and isokinetic contractions." Journal of ELectromyography and Kinesiology **10**: 425-431.

Kernell, D., Monster, A.W. (1982). "An intracellular study of gastrocnemius motoneurones of the cat." Exp Brain Res **46**: 197-204.

Khodiguian, N., Cornwell, A., Lares, E., DiCaprio, P.A., and Hawkins, S.A (2003). "Expression of the bilateral deficit during reflexively evoked contractions." J Appl Physiol **94**(1): 171-178.

Khuvasanont, T. (2002). Age-related ankle strength degradation and effects on slip-induced falls, Virginia Polytechnic Institute and State University **M.S.**

Kilbom, A. (1990). Measurememnt and assessment of dynamic work Bristol, PA, Taylor and Francis.

Kim, S., Lockhart, T., Yoon, H.Y. (2005). "Relationship between age related gait adaptations and required coefficient of friction." Safety Science **43**: 425-436.

Komi, P. V., Tesch, P. (1979). "EMG frequency spectrum, muscle structure and fatigue during dynamic contractions in man." Eur J Appl Physiol **42**: 41-50.

Larsson, L., Grimby, G., Karlsson, J. (1979). "Muscle strength and speed of movement in relation to age and muscle morphology." Journal of Applied Physiology **46**: 451-

456.

Lattanzio, P. J., Petrella, R.J., Sproule, J.R., Fowler, P.J. (1997). "Effect of fatigue on knee proprioception." Clin J Sport Med(7): 22-27.

Layne, L. A., Pollack, K. M. (2004). "Nonfatal occupational injuries from slips , trips and falls among older workers treated in hospital emergency departments, United States 1998." Am J Ind Med. **46**(1): 32-41.

Leamon, T. B., Murphy, P.L. (1995). "Occupational slips and falls: more than a trivial problem." Ergonomics **38**(3): 487-498.

Lin, L. J., Chiou, F. T., & Cohen, H.H. (1995). "Slip and Fall accident prevention- A review of research, practice, and regulations." Journal of safety research **26**: 203-212.

Lindsrom, L., Magnusson,R., Petersen, I. (1970). "Muscular fatigue and action potential conduction velocity changes studied with frequency analysis of EMG signals." Electromyography **4**: 341-353.

Lipscomb, H. J., Glazner, J. E., Bondy, J., Guarini, K., Lezotte, D. (2006). "Injuries from slips and trips in construction." Applied Ergonomics **37**: 267-274.

Liu, J., Lockhart, T.E. (2006). "Comparison of 3D joint moments using local and global inverse dynamics approaches among three different age groups." Gait & Posture **23**: 480-485.

Lockhart, T. E., Wolstad, J., & Hsiang, S.M (2000). Predictions of falls using a robust definition of slip distance and adjusted required coefficient of friction. Proceedings of IEA/HFES Conference.

Lockhart, T. E., Wolstad, J., Smith, J. L., Ramsey, J. D. (2002). "Effects of age related sensory degradation on perception of floor slipperiness and associated slip parameters." Safety Science **40**: 689-703.

Lockhart, T. E., Wolstad, J., Smith, J. L. (2003). "Effects of age-related gait changes on biomechanics of slips and falls." Ergonomics **46**(12): 1136-1140.

Lockhart, T. E., Wolstad, J. C., & Smith, J. L. (2002a). "Assessment of slip severity among different age groups." Metrology of pedestrian locomotion and slip resistance **ASTM STP 1424**.

Lockhart, T. E., Wolstad, J., & Hsiang, S.M (2005). Effects of Aging on the

Biomechanics of Slips and Falls. Proceedings of HFES Conference.

Maiti, J., Bhatacharjee, A., Bangdiwala, S.I. (2001). "Loglinear model for analysis of cross-tabulated coal mine injury data." Injury control and safety promotion **8**: 229-236.

Maki, B. E., Mclloy, W.E., Perry, S.D. (1996). "Influence of lateral destabilization on compensatory stepping response " Journal of Biomechanics **29**(343-353).

Manning, D. P., Ayers, I., Jones, C.,Bruce, M. & Cohen, K. (1988). "The incidence of underfoot accidents during 1985 in a working population of 10,000 Merseyside people." Journal of Occupational Accidents **10**: 121-130.

Masuda, K., Masuda, T., Sadoyama, T., Inaki, M., Katsuta, S (1999). "Changes in surface EMG parameters during static and dynamic fatiguing contractions." J Electromyogr Kinesiol **9**: 39-46.

Miller, P. K., Bird, A. M (1976). "Localized muscle fatigue and dynamic balance." Percept Mot Skills **42**: 135-8.

Miller, R., Kent-Braun, JA., Sharma, KR., WEiner, MW. (1995). "Mechanisms of human muscle fatigue: quantitating the contribution of metabolic factors and activation impairment." Adv Exp Med Biol. **384**: 195-210.

Mills, P. M., Barrett, R.S. (2001). "Swing phase mechanics of healthy young and elderly men." Hum. Move. Sci. **20**: 427-446.

Mirka, A., Black, F.O. (1990). "Clinical application of dynamic posturography for evaluating sensory integration and vestibular dysfunction." Neurological Clinics **8**(2): 351-359.

Mizrahi, J., Verbitsky, O., Isakov, E. (2000). "Fatigue-related loading imbalance on the shank in running: a possible factor in stress fractures." Annals of Biomedical Engineering **28**: 463-469.

Mizrahi, J., Verbitsky,O., Isakov,E. (2001). "Fatigue- Induced changes in decline running." Clinical Biomechanics **16**: 207-212.

Moyer, B. E., Chambers, A.J.,Redfern, M.S., Cham, R. (2006). "Gait parameters as predictors of slip severity in younger and older adults." Ergonomics **4**: 329-343.

Murray, C. J. L., Lopez,A.D., (1996). "Global and regional descriptive epidemiology of disability:incidence,prevalence,health,expectancies, and years lived with disability." The Global Burden of Disease. Harvard University Press, Boston: 201-246.

Nardone, A., Tarantola, J., Giordano, A., Schieppati, M. (1997). "Fatigue effects on body balance." Electroencephalogr Clin Neurophysiol **105**: 309-320.

Nashner, L. M. (1980). "Balance adjustments of human perturbed while walking." Journal of Neurophysiology **44**: 650-664.

Nashner, L. M., Peters, J.F. (1990). "Dynamic posturography in the dignosis and management of dizziness and balance disorders." Neurological Clinics **8**: 331-349.

Neter, J., Kutner, M.H., Nachtssheim, C.J., and Wasserman, W. (1996). Applied linear statistical model. Irwin, McGraw-Hill/.

Nicol, C., Kmoi, P.V, Horita, T.,Kyrolainen,H., Takala, T.E.S (1996). "Reduced stretch-reflex sensitivity after exhausting stretch-shortening cycle." Eur J Appl Physiol **72**: 401-409.

NIOSH (2000). A summary of Surveillance Findings and Investigative Case Reports,NIOSH(DHHS). National Traumatic Occupational Fatalities surveillance system**:** Publication No. 2000-116.

Nussbaum, M. A. (2001). "Static and dynamic myoelectric measures of shoulder muscle fatigue during dynamic intermittent exertions of low to moderate intensity." Eur J Appl Physiol **85**: 299-309.

Oberg, T., Sandsjo, L., and Kedefors, R. (1994). "Subjective and objective evaluation of shoulder muscle fatigue." Ergonomics **37**(8): 1323-1333.

Pai, Y. C., Patton, J. L. (1997). "Center of mass velocity - position predictions for balance control." J Biomech. **30**: 347-354.

Perkins, P. J. (1978). "Measurement of slip between the shoe and ground during walking" American Society of Testing and Materials, Special Technical Publication **649**: 71-87.

Perkins, P. J., Wilson, M. P. (1983). "Slip resistance testing of shoes - New developments." Ergonomics **26**: 73-82.

Perry, J. (1992). Gait analysis: Normal and Pathological function, New York: Mc Graw-Hill, Inc.

Redfern, M. S., and Bidanda, B. (1994). "Slip resistance of the shoe floor interface under biomechanically relevant conditions." Ergonomics **37**(3): 511-524.

Redfern, M. S., and DiPasquale. (1997). "Biomechanics of descending ramps." Gait & Posture **6**: 119-125.

Robinovitch, S. N., Heller, B., Lui, A., Cortez, J (2002). "Effect of strength and speed of torque development on balance recovery with the ankle strategy." J Neurophysiol **88**: 613-620.

Saggini, R., Pizzigallo, E., Vecchiet, J. (1998). "Alteration of spatial-temporal parameters of gait in chronic fatigue syndrome patients." J Neurol Sci **154**: 18-25.

Sejersted, O. M., Vollestad, N.K. (1993). Physiology of muscle fatigue and associated pain. In progress in Fibromyalgia and Myofascial Pain, Elsevier Science Publishers.

Seliga, R., Bhattacharya, A., Succop, P., Wickstrom, R., Smith, D., Willeke, K. (1991). "Effect of work load and respirator wear on postural stability, heart rate, and perceived exertion." AM. IND. HYG. ASSOC. J. **52**(10): 417-422.

Skinner, H. B., Wyatt, M.P, Hodgdon, J.A., Conard, D. W., Barrack, R.L. (1986). "Effect of fatigue on joint positioning of the knee." J Orthop Res **4**: 112-118.

Slobunov, S., Hallet, M.,Stanhope, S,and Shibasaki, H (2005). "Role of cerebral cortex in human postural control : an EEG study." Clinical Neurophysiology **116**: 315-323.

Smith, G. S., and Veazie, M. A. (1998). Principles of prevention: the public health approach to reducing injuries in the workplace. Encyclopedia of Occupational Health and Safety, Geneva. Switzerland: International Labor Office.

Sparto, P. J., Parnianpour, M., Reinsel, T.E., Simon, S. (1997). "The effect of fatigue on multijoint kinematics, coordination, and postural sway stability during a repetitive lifting task" J Orthop Sports Phys Ther **25**(1): 3-12.

Sun, J., Walters, M., Svensson, N., Lloyd, D. (1996). "The influence of surface slope on human gait characteristics: a study of urban pedestrians walking on an inclined surface." Ergonomics **39**: 677-692.

Svantesson, U., Osterberg, U., Thomee, R., Peeters, M., Grimby, G. (1998). "Fatigue during repeated eccentric concentric and pure concentric muscle actions of the plantar flexors." Clinical Biomechanics **13**: 336-343.

Swaen, G. M. H., van Amelsvoort, L.G., Bultmann, U., Kant, I. J. (2003). "Fatigue as a risk factor for being injured in an occupational accident: results from the Maastricht Cohort Study." Occup Environ Med **60**(Suppl I): i88-i92.

Syed, I. Y., Davis, B.L. (2000). "Obesity and osteoarthiritis of the knee: hypotheses concerning the relationship between ground reaction forces and quadriceps fatigue in long-duration walking." Medical Hypotheses **54**(2): 182-185.

Tideiksaar, R. (1997). Falling in Old Age., Springer Publishing Company,Inc.,New York.

Tisserand, M. (1985). "Progress in the prevention of falls caused by slipping." Ergonomics **28**(7): 1027-1042.

Troop, H., Odenrick, P. (1988). "Postural control in single-limb stance." J Orthop Res **6**: 833-839.

Vandervoort, A. A., Sale, D.G., and Moroz, J. (1984). "Comparison of motor unit activation during unilateral and bilateral leg extension." J Appl Physiol **56**: 46-51.

Vollestad, N. (1997). "Measurement of human muscle fatigue." J Neurosci Methods **74**: 219-227.

Voloshin, S. A., Mizrahi, J.,Verbitsky, Oleg., Isakov, Eli. (1998). "Dynamic loading on the human musculoskeletal system- effect of fatigue." Journal of Clinical Biomechanics **13**: 515-520.

Wilson, S. E., Granata, K. P. (2000). Active knee stiffness decreases after fatigue. 24th Annual meeting of the American Society of Biomechanics, Chicago.

Winter, D. A. (1983). "Moments of force and mechanical power in jogging." Journal of Biomechanics **16**: 91-97.

Winter, D. A. (1991). The Biomechanics and motor control of Human Gait: Normal, Elderly and Pathological., Waterloo Biomechanics, Graphic Services.

Wolfson, L., Judge, J., Whipple, R., King, M. (1995). "Stength is a major factor in balance, gait, and the occurrence of falls." J Gerontol **50A**: 64-67.

Yaggie, J. A. M., S.J. (2002). "Effects of isokinetic ankle fatigue on the maintenance of balance and postural limits." Archival of Physical Medicine and Rehabilitation **83**: 224-228.

Yoshino, K., Motoshige, T., Araki, T., Matsouka, K. (2004). "Effect of prolonged free-walking fatigue on gait and physiological rhythm." Journal of Biomechanics **37**: 1271-1280.

APPENDIX I

Informed Consent Form

Grado Department of Industrial and Systems Engineering
Virginia Tech

TITLE: Biomechanical Analysis of Slip-Induced Falls

PRINCIPAL INVESTIGATOR: Thurmon E. Lockhart Ph.D.

PURPOSE

This is an experiment to investigate the changes in biomechanical parameters and ground reaction forces during fatigue. The objective of this experiment is to measure the fatigue effect on different surface conditions with or without contaminant (soapy water).

PROCEDURE

The study will last two days with the first day consisting of a familiarization session and body composition measurements, and the second day consisting of 20 minute walking experiment. Prior to this experiment, you will be given an opportunity to walk around the laboratory wearing a harness to familiarize yourself with the equipment (fall arresting rig), and floor surfaces. On the second day, you will be asked to walk across the soapy or dry floor surface for 20 minutes. While you are walking along the path, please keep your eyes looking straight ahead and try to maintain the speed that you practiced.

After the familiarization session, you will be asked to walk on specially prepared floor surfaces. The floor surface which you will walk across may or may not be slippery. As you experienced in the familiarization session, the harness system will protect you if you slip and begin to fall. The fall arresting rig will stop the motion of the tracking device and allow you to "fall or slip" only 3 or 4 inches.

Additionally, strength test will be performed using a dynamometer. To test the leg strength, you will be asked to sit on Biodex chair with backrest. The vertical height of the chair will be adjusted to accommodate your popliteal height. The speed selector will be set at 30, 60 and 90 degrees per second. The tests will be performed for the range of 0 to 90° of flexion. The leg would begin in the neutral, 0° position corresponding to a straight, extended leg, and proceed to trace out a bending motion until the upper and lower portions of the leg formed a 90° angle. You will be asked to apply three maximal torques at each velocity as suggested by the Biodex Dynamometer Exercise Manual.

RISKS OF PARTICIPATION

Minor muscle sprain, if you lose my balance while walking on the floors and due to Biodex exertion.

BENEFITS and COMPENSATION

The benefits to you are a better understanding of floor surface slipperiness which could lead to preventing slips and falls in the elderly. Additionally, monetary compensation will be provided ($10.00 per hour).

ANOYNMITY AND CONFIDENTIALITY

The data from this study will be kept strictly confidential. No data will be released to anyone but the principal investigator and graduate students involved in the project without written consent of the subject. Data will be identified by subject number.

FREEDOM TO WITHDRAW

You are free to withdraw at any time from the study for any reason. Circumstances may come up that the researcher will determine that you should not continue as a subject in the study. For example, an illness could be a reason to have the researchers stop your participation in the study.

APPROVAL OF RESEARCH

This research has been approved, as required, by the Institutional Review Board for Research Involving Human Subjects at Virginia Tech, and by the Grado Department of Industrial and Systems Engineering. You will receive a copy of this form to take with you.

SUBJECT PERMISSION

I have read the informed consent and fully understand the procedures and conditions of the project. I have had all my questions answered, and I hereby give my voluntary consent to be a participant in this research study. I agree to abide by the rules of the project. I understand that I may withdraw from the study at any time.

If I have questions, I will contact:

Principal Investigator: Thurmon E. Lockhart, Assistant Professor, Grado Department of Industrial and Systems Engineering, 231-9088.

Chairman, Institutional Review Board for Research Involving Human Subjects: David Moore, 231-4991.

Signature of Subject ____ Date:

Signature of Project Director or his Authorized Representative:
____ Date:

Signature of Witness to Oral Presentation:
____ Date:

APPENDIX II

Personal Data and Medical History

Virginia Tech, ISE Department

Biomechanical Analysis of Slip-Induced Falls

Date _____________

Name ____ Age _____________
Sex __________ Height (ft in) _______________ Weight (lb) _____________
In case of emergency contact: Name ______________________ Phone ____________

Check if susceptible to:
Shortness of breath _____ Dizziness _____ Headaches _____ Fatigue _____
Pain in arm, shoulder or chest ______
If you checked any of the items above, please explain: ___________________________
____

Have you ever had a heart attack ? ______ If so, give history:
____

Are you currently taking any type of medication ? ______ If so, please explain:
____

Have you had or do you now have any problems with your blood pressure ? _______ If so, please explain: ____

In the last 6 month, have you had any back pain ? ______ If so, please explain:
____

Have you had or do you now have a hernia ? ______ corrective date:

Have you had or do you have any ankle, knee or hip problems ? ______ If so, please explain:____

Have you had any surgical hip or knee replacements (i.e., ACL replacement)? ______ If so, please explain: ____

Have you had any other surgical procedure performed on ankle, knee or hip? ______ If so, please explain: ____

Have you currently have osteoporosis or treated with osteoporosis? If so, please explain:
____

Have you had or do you now have any inner ear or balance problems? ________ If so, please explain: ____

Have you ever experienced slips and falls ? _______ If so, how long ago : ___________

VITA

PRAKRITI PARIJAT

⌧ - 305 Hunt Club Road, Apt 6600C, Blacksburg, VA -24060
💻 – pparijat@vt.edu ☎ – (540) 250 1463

Objective

To obtain a full time position that provides professional and research experience in an environment characterized by continuous learning and challenges in the field of Biomedical Engineering and rehabilitation.

Education Qualifications

MS, *Biomedical engineering and sciences* (Biomechanics option), Virginia Tech, Blacksburg (Fall 2004-expected June 2006).	**GPA 3.6/4.0**
BE, *Biomedical Instrumentation*, Avinashilingam Deemed University, Tamilnadu, India (Fall 1998-Summer 2002).	**GPA 8.9/10**

Graduate Courses

Mammalian Physiology	Human physical capabilities	Introduction to Biomedical engg.
Advanced topics in Human Factors Engineering	Engineering analysis of physiological systems	Musculoskeletal Biomechanics
Impact Biomechanics	Statistics in Research Design	Biodynamics and control

Professional Experience

- **Graduate Research Assistant** , Locomotion research lab, from Jan 2006 – till date, Virginia Tech.
- **Graduate Teaching Assistant** for the course **Human physical capabilities** from August 2005- Dec 2005, Virginia Tech.
- **Graduate Research Assistant** , Locomotion research lab, from July 2004 – May 2005, Virginia Tech.
- **Research Assistant**, Center for Biomedical Engineering, from Dec 2003 – April 2004, Indian Institute Of Technology, Delhi, India.
- Internship with “GE Medical Systems, Bangalore, India” from May 2001- June 2001 for Installation of PACS (Picture archival and communication systems) in collaboration with Fortis Heart Institute, Mohali, India.

Technical Skills

Programming: C, Matlab, and LabVIEW
Measurement Techniques: EMG, Force platforms, Gait analysis software (Qualysis).

Publications

- Effect of Load Carrying on Trunk dynamics during unexpected slip and recovery, Parijat.P, Lockhart T et al., Proceedings of the XIX Annual International Occupational and safety Conference, 2005, Las Vegas, Nevada.
- EEG Characterization and Classification for ALS Patients, Poster Presentation, Parijat P, Sahu S, International Conference on Brain Research 2003, National Brain Research Center, Delhi, India.
- Understanding Pacemakers, Parijat P, Chacko S , Proceedings of the IETE National Symposium 2000, Tamilnadu, India.
- Load Carrying effects on Low back moment and trunk dynamics, Parijat P, Liu J, Lockhart T, Poster session, 4th graduate student research symposium 2005, Virginia Tech.

Projects

- Evaluation of effects of load carrying on trunk kinematics and low back moment generation. Project involved extensive human subject data collection , kinematic and kinetic analysis, and report generation at locomotion research lab, Virginia Tech, 2005.
- Design and Development of non-invasive blood flow measurement system for classification of brain states as relax and planning. The project consisted of hardware designing using LabVIEW 6.1 and signal analysis using Matlab tools,2003-2004, IIT, Delhi, India.

- Testing and ergonomic evaluation of Flex-o-chair designed for routine exercise for the elderly. Project involved comparison of EMG activity and aerobic capacity while walking and while exercising on the chair, locomotion research lab, Virginia Tech, 2004.
- Development and testing of automated –fetal transducer unit, funded by “GE Medical Systems" Bangalore, India, 2001- 2002. Project involved hardware designing and interfacing with already existing software tools.

Honors and Affiliations

- Research Fellowship, Indian Institute Of Technology, Delhi, India (May 2003- March 2004)
- Vice-president to the Virginia Tech student chapter of ”Biomedical Engineering society”, member of “Human Factors and Ergonomic society (HFES)”
- Rank Holder (2nd) throughout the junior and senior years in BE, Biomedical Instrumentation Engineering (2000-2002).
- Recipient of outstanding upcoming player at the national level ball badminton championship, chandigarh, India in 1997.