Central Nervous System Control of Dynamic Stability during Locomotion in Complex Environments

By

Michael James MacLellan

A thesis
presented to the University of Waterloo
in fulfillment of the
thesis requirement for the degree of
Master of Science
In
Kinesiology

Waterloo, Ontario, Canada, 2006

Author’s Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

A major function of the central nervous system (CNS) during locomotion is the ability to maintain dynamic stability during threats to balance. The CNS uses reactive, predictive, and anticipatory mechanisms in order to accomplish this. Previously, stability has been estimated using single measures. Since the entire body works as a system, dynamic stability should be examined by integrating kinematic, kinetic, and electromyographical measures of the whole body. This thesis examines three threats to stability (recovery from a frontal plane surface translation, stepping onto and walking on a compliant surface, and obstacle clearance on a compliant surface). These threats to stability would enable a full body stability analysis for reactive, predictive, and anticipatory CNS control mechanisms. From the results in this study, observing various biomechanical variables provides a more precise evaluation of dynamic stability and how it is achieved. Observations showed that different methods of increasing stability (eg. Lowering full body COM, increasing step width) were controlled by differing CNS mechanisms during a task. This provides evidence that a single measure cannot determine dynamic stability during a locomotion task and the body must be observed entirely to determine methods used in the maintenance of dynamic stability.

Acknowledgements

I would like to thank Dr. Aftab Patla for challenging me as a student and a person during the course of my studies thus far at the University of Waterloo. The help of Dr. James Frank and Dr. Richard Staines is appreciated for guiding me during the process of my Masters thesis. Finally, I would like to thank all faculty and students working in the Gait and Posture Laboratory for their support. Funding for this research was provided by NSERC and an OGS Scholarship.

Dedication

This work is dedicated to my fiancée Catherine Allin and my family Jim, Cecilia, and Patrick MacLellan. Without the love and support from these people, my academic endeavors would not be possible. I am forever grateful for having all of you in my life.

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1.1.1 - Introduction

During locomotion, stability is constantly challenged due to full body center of mass (COM) being controlled within a continually moving base of support (BOS) (Patla 2003). To maintain stability during locomotion, the central nervous system (CNS) employs reactive, predictive and anticipatory control mechanisms (Patla 2003). Reactive control relies on sensory detection of a perturbation in response to threats of stability. During predictive control, the central nervous system makes estimations from past experiences to proactively respond to threats to stability by voluntary movements. Anticipatory control uses visual feedback to possibly avoid or prepare for threats to stability. This thesis explores the mechanisms used by the CNS in order to facilitate particular considerations needed to maintain reactive, predictive and anticipatory control of dynamic stability. Previously, research has examined stability by observing relationships between COM position and BOS position. Recently, research has suggested that not only should the COM position be related to BOS position, but COM velocity within the BOS (Hof et al. 2005). The studies in this thesis examine the differing CNS mechanisms used in controlling locomotion by extensively examining new determinants of dynamic stability in order to infer how the CNS maximizes balance during instances when stability is threatened.

1.1.2 - Reactive Control of Locomotion

During the event of an unexpected perturbation, the primary means of regaining stability are through reactive mechanisms. The sensory systems used in the maintenance of balance include vision, vestibular, and kinesthetic systems. In the event of an unexpected perturbation, reactions used to regain stability are quick and therefore, vision may not be the first line of defense used because a reaction to visual stimulus takes approximately 200 ms (Oates et al. 2005). Instead, vestibular and kinesthetic inputs are used. The kinesthetic system, for example, senses stretching of muscles which occurs during slipping events and reflexes are elicited to maintain stability. Reactive control of locomotion is studied largely by perturbation experiments. In these studies, an unknown threat to stability is elicited and responses are observed. In the late 1970's, Lewis Nashner initiated a group of studies in which he observed postural changes in response to a rapidly moving platform. Nashner's early studies involved eliciting a postural response using rapid anterior and posterior translations to examine electromyography (EMG) changes and to determine how the nervous system is involved in making these changes. The initial EMG responses occurred at 80-90 ms and it was determined that these reactive responses to unknown perturbations must be motor programs due to timing and stereotypical responses (Nashner 1982). These studies are still presently used with kinematics and kinetics observed along with EMG patterns. These are the primary mechanisms used in maintaining stability following an unexpected perturbation.

1.1.3 - Predictive Control of Locomotion

In predictive control of locomotion, feed-forward control is utilized by the CNS to maximize stability during a threat to locomotion. For example, when a threat to stability is initially introduced to the CNS, the response differs from subsequent threats to stability. Results have shown that the CNS may evoke a “startle response” during an initial perturbation which subsides in following perturbations (Marigold & Patla 2002; Oates et al. 2005). The responses following the startle response are subsided through predictive control of the CNS. This startle response was documented previously in a study in which an acoustic sound was given to participants while walking on a treadmill (Nieuwenhuijzen et al. 2000). This startle response from the auditory signal subsided after two presentations (Nieuwenhuijzen et al. 2000). During slipping on a set of steel rollers, Marigold and Patla (2002) showed that during the initial slip, heightened muscle activity, increased breaking impulse, increased ankle dorsiflexion at heel contact, and a decreased vertical COM. The CNS employed results from prior experiences to decrease muscle activity, decrease braking impulse, land with more of a flat foot, and increase vertical center of mass (Marigold & Patla 2002). These changes occurred after the first slip recovery (Marigold & Patla 2002). This evidence shows how the CNS uses feed-forward control to manipulate various aspects of locomotion to maintain dynamic stability during repeated perturbations.

1.1.4 - Anticipatory Control of Locomotion

Anticipatory control of locomotion is used to change stability in preparation of a threat to stability. Modifications in gait patterns have been examined when anticipating a slippery surface (Cham & Redfern 2002) and when knowledge is given that a slip will occur on a set of steel rollers (Marigold & Patla 2002). Cham and Redfern (2002) examined how gait was altered if a possibility existed that a surface may be slippery. When anticipating a slippery surface, ground reaction forces and ankle velocity decreased at heel contact (Cham & Redfern 2002). When a slippery surface was anticipated on an incline, Cham and Redfern (2002) observed decreases in stance duration and step length when stepping on the slippery surface. It was suggested that these modifications in gait occur by decreases in hip, knee, and ankle joint moments during anticipation (Cham & Redfern 2002). Similar results were observed by Marigold and Patla (2002) in which a reduced braking impulse and rate of loading occurred along with a shift of medial-lateral COM towards the stance limb and a flat foot landing. When on the slippery surface, participants used a “surfing strategy” in which arms were raised forward and outward while traveling on the surface (Marigold & Patla 2002). These adaptations would in turn increase stability when stepping on a slippery surface. These results indicate a more cautious gait strategy is used when anticipating a step onto a slippery surface. This shows how the CNS can change gait patterns in order to anticipate a threat to stability.

1.1.5 - Determinants of Stability

The study of the mechanics involved with stability is important because by determining what is needed to maintain balance, one is able to determine what conditions increase the risk of falling. In previous studies, stability has been determined by calculating the distance from the COM to the BOS (Shumway-Cook and Woolacott 1995, Winter 1995). The theory behind this being that COM movement is directly controlled by application of force by the foot on the ground (center of pressure). This force on the ground would create a moment on the COM and would therefore accelerate (Winter 1995). This definition of stability would be appropriate during stable stance, but during locomotion, additional factors exist in the maintenance of stability. During locomotion, the COM is constantly moving, and therefore location of COM alone is insufficient in determining stability. It was suggested by Pai and Patton (1997) that due to the movement of the COM, the velocity would also be taken into account when determining stability. In theory, during locomotion the position of the COM may be outside of the BOS but the COM may be moving towards the BOS, which would create a stable situation. It can be seen that COM velocity is an important determinant of dynamic stability.

The idea that COM velocity should be taken into account into dynamic stability situations was extended by Hof et al. (2005) by applying the inverted pendulum model of stance to COM velocity. Hof et al. suggested to multiply a factor of $(l/g)^{1/2}$ (l = leg length, g = acceleration due to gravity) to the velocity of the COM to take into account physics due to the inverted pendulum model of stability. This model along with simple COM location and COM velocity calculations are used to determine stability in these studies.

1.1.6 - Rationale of Thesis

Previous research in locomotion has examined stability using single measures and stability is predicted. Many studies do not incorporate the numerous biomechanical measurements taken during a study to determine how the CNS functions to maximize stability. This type of analysis may not be appropriate because the body functions as an entity to arrive at a goal. This thesis will integrate kinematic, kinetic, and EMG measures to analyze how the CNS controls full body movement in order to maximize stability.

As stated, the CNS controls locomotion through reactive, predictive, and anticipatory mechanisms. This group of studies will examine these control mechanisms and incorporate various biomechanical measures in order to determine how the CNS maximizes stability when a threat is encountered. The initial study will examine reactive control of locomotion by observing how stability is regained following a medio-lateral (M/L) surface translation. Predictive control of stability will be examined by observing how the CNS adapts to stepping onto and walking on a compliant surface. The final study will observe how competing demands of maintaining stability and crossing an obstacle are met while walking on a compliant surface to examine anticipatory control of stability.

In this group of studies, it is hypothesized that the CNS controls various aspects of human movement in order to maintain stability. This will be shown by controlling aspects such as step characteristics, COM position and velocity, muscle activations, and joint kinetics. All of these characteristics of movement will be controlled in order to regulate COM relationships with the BOS. It will be seen that dynamic stability cannot be determined by a single measure but by incorporating all aspects of movement together to meet a common goal of maintaining stability.

2.1.1 – Active Control by Central Nervous System of Dynamic Stability during Frontal Plane Surface Translations

Studies involving surface translations and moving platforms have been used to examine how the central nervous system (CNS) maintains stability during perturbations that simulate real life experiences. Situations such as this can occur in accelerating or decelerating buses and subways and can somewhat simulate a slipping event. To date surface translations have been examined during stable stance and walking and from these studies, inferences on how the CNS maintains stability have been made.

Surface translations can pose a problem to stability during everyday life. To date, much research has examined surface translations during stable stance and some have examined surface translations and slips during locomotion. Early studies of surface translations were performed by Nashner (1982) where translations were elicited in forward and backward directions during stable stance. It was concluded that due to the speed and repeatability of postural corrections, the CNS must use motor programs to maintain stability. It was also observed that muscle activations were organized in a distal to proximal order (Nashner, 1982) and this was repeated in Henry, Fung, & Horak (1998b) during frontal plane stable stance surface translations. Henry, Fung, & Horak (1998a) also showed that body segment kinematics followed a distal to proximal sequence where the trunk was the last segment to make and complete corrections. It was suggested from these results that recovery from standing anterior / posterior perturbations are similar to lateral perturbations and differences only exist due to differing biomechanical constraints that occur in these planes (Henry, Fung, & Horak, 1998a). These studies are limited in the fact that dynamic stability cannot be examined under this paradigm.

Research examining surface translations and slips during locomotion has only examined perturbations in the saggital plane. It has been observed that during anterior surface translations during locomotion, leg musculature activation follows a distal to proximal sequence (Tang et al. 1998). During an anterior slip on a set of rollers, this distal to proximal sequencing was not observed (Marigold and Patla, 2002). Marigold and Patla (2002) concluded muscle activation strategies were employed to maximize stability. One strategy suggested was a limb flexor which would serve to lower full body center of mass (COM) (Marigold and Patla, 2002). You et al. (2001) also observed an increase in center of mass – base of support (COM-BOS) difference that would increase stability. From these results, the CNS employs mechanisms to increase stability in instances of slips and surface translations in the saggital plane.

Many locomotion studies have only examined slips and surface translations in the sagittal plane, and two have examined surface translations at 45 degrees from the sagittal plane (Oddsson et al. 2004; Wall et al. 2002). A limited analysis is provided in these studies because the main observations were made on COM and foot placements. Perturbations in the frontal plane (medial/ lateral) have never been examined previously. To address the issues stated, this study will consist of a thorough examination of surface translations in which pure medial and lateral perturbations will be elicited at different times in stance. Since differing perturbation times have not been examined during surface translations, it is unknown if CNS responses will change due to this timing. Observing both medial and lateral surface translations would be interesting due to the effects they would have on the base of support (BOS). A lateral perturbation would widen the BOS while a medial perturbation would create a narrow BOS. This would have tremendous implications on COM position to maintain stability. It has been suggested that frontal plane stability is controlled mostly by the hip musculature while sagittal plane stability is controlled by ankle musculature (Winter et al. 1996). The different biomechanical constraints seen in these planes would create different recovery strategies than previously observed. To address the issue of timing, these medial and lateral perturbations will be elicited at two different times: heel contact and 200 ms following heel contact (delayed). Due to the differences in the relationship between full body COM and BOS during these times, recovery strategies should differ between these conditions.

During this study, it is hypothesized that the CNS will employ mechanisms to maximize stability during instances of frontal plane perturbations. Previous research has illustrated that muscle activation follows a distal to proximal sequence. Since the hip musculature will be stretched in instances of perturbation, it is thought the hip musculature will be the first to respond to the perturbation. In the perturbed limb, it is hypothesized that hip adductors will be activated in a lateral perturbation while hip abductors will be activated during a medial perturbation. These muscle activations will stabilize the pelvis during the perturbations. The COM will most likely be displaced the furthest during heel contact perturbations because when perturbations occur later in stance, the CNS may not have sufficient time to respond prior to contralateral heel contact. It is hypothesized that the steps following the perturbation will be wider than baseline steps to increase base of support following the perturbation in order to regain stability.

2.2.1 - Methods

2.2.2 - Participants

Eight young adults (4 female, 4 female, age 23.8 +/- 1.16 years, 80.8 +/- 15.15 kg, 177.9 +/- 8.54 cm) from the University of Waterloo participated in this study. To be eligible for this study, participants reported no previous muscular, joint, balance, or neuromuscular problems. This study was approved through the Office of Research Ethics at the University of Waterloo and each subject signed a consent form prior to participation.

2.2.3 - Sliding Platform

The sliding platform consisted of an Advanced Medical Technology Inc. (AMTI) force plate (0.51 by 0.465 meters) which is attached to a 1.64 meter track below the surface of the floor. A computer controlled stepper motor caused a 0.15 meter translational movement of the platform at a velocity of 0.3 m/s. This sliding platform was flush with the floor as to not make a gap between the floor and the platform.

2.2.4 - Protocol

All participants reported to the laboratory at which time they were outfitted with markers for motion tracking and electrodes for muscle activity recording. Perturbations were elicited to the right foot by the sliding platform in a medial or lateral direction with respect to the midline of the body. Timing of perturbations occurred at approximately heel contact and 200ms following heel contact. These were termed heel contact and delayed perturbations. The study consisted of 160 randomized trials with 40 perturbation trials which were randomly allocated. These 40 perturbations were divided into 20 medial and 20 lateral perturbations. These 20 perturbations were further divided into 10 heel contact perturbations and 10 delayed perturbations. The remaining 120 trials were baseline walkthrough trials.

The participants were given instructions prior to the study to start and stop walking when given a signal by the experimenter and to continue walking as normal as possible throughout the entire trial. At the start of each trial, participants were asked to begin walking. Participants started walking at a position approximately 5 steps prior to stepping on the sliding platform and were situated such that a right foot fall occurs on the platform. The participants walked until the experimenter informed the participant to stop walking, approximately 6 steps after the participant stepped on the sliding platform. All steps were named with respect to heel contact on the moving platform (N-1, N, N+1, N+2, and N+3) (Figure 1A).

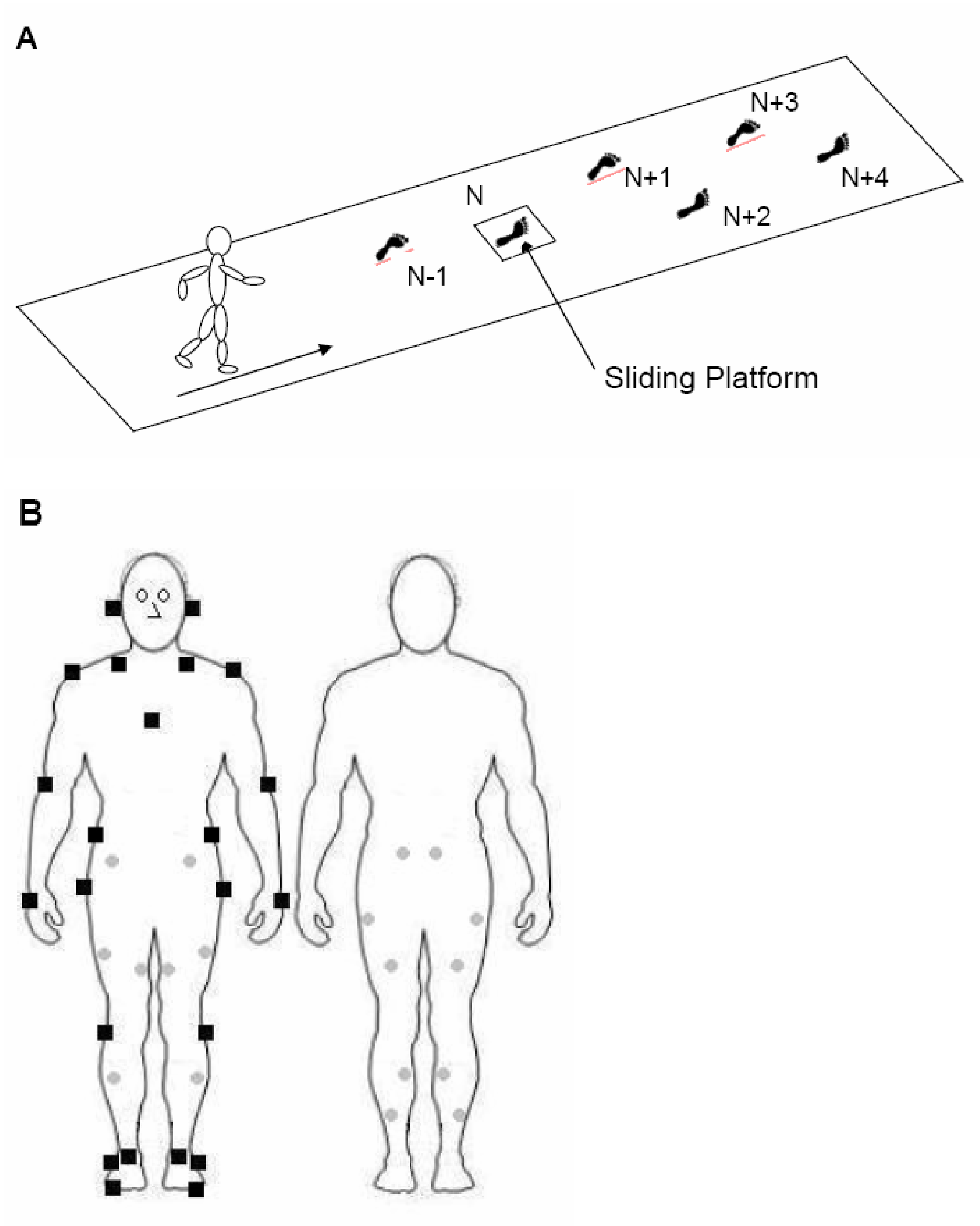


Figure 1: Experimental Setup. A: Organization of sliding force plate and naming of steps. B: Location of iREDs (black squares) and EMG electrodes (grey circles).

2.2.5 - Kinematics

Kinematic data was obtained through a three camera three-dimensional motion capture system (Optotrak 3020, Northern Digital Inc., Waterloo, Ontario, Canada). The motion capture system was sampled at 60 Hz. Infrared emitting diodes (iREDS) which track movement for the Optotrak system were placed bilaterally on the fifth metatarsal, lateral malleolus, anterior aspect of ankle, fibular head, greater trochanter, anterior superior iliac spine, mid-clavicle, acromioclavicular joint, olecranon, radial styloid, ear, and xiphoid process (Figure 1B). Trunk three-dimensional angles were determined using Cardan methods and 2.5 dimensional lower limb joint angles will be determined using methods outlined in Winter (2005). All kinematic data was low pass filtered at 7 Hz using a dual-pass 2nd order Butterworth filter. A video camera was set up as a visual log of each participant.

2.2.6 - Electromyography

Muscle activation was measured by surface electromyography (EMG) (Bortec, Calgary, Canada) from 16 lower limb muscles (bilateral: rectus abdominis oblique, lower erector spinae, rectus femoris, biceps femoris, gluteus medius, adductor longus, tibialis anterior, and gastrocnemius) using Ag-AgCl (Kendall Medi-Trace, Chicopee, MA, USA) electrodes (Figure 1B). Skin was prepared as proposed in Hermens et al. 2000. Signals were band pass filtered at

10-1000 Hz and sampled at 1200 Hz (to correspond with Optotrak sampling). The common mode rejection of the EMG system was 115 dB with input impedance of 10 GOhms.

2.2.7 - Data Analysis

Kinematic data was used to determine full body COM, posture COM (body COM minus perturbed leg COM), perturbed leg COM, three-dimensional trunk angles, foot placement, and stability. An 11-linked segment model (Winter 2005) was used to determine full body COM. To determine three-dimensional trunk angles, a coordinate system was setup during the standing calibration trial. From the COM and trunk segment angle data, baseline walkthrough trials for each participant were subtracted from perturbation trials and full body COM positions and trunk angles were determined at each step. Differences in posture COM and leg COM between step N and N+1 were examined by multiplying the change in medial / lateral (M/L) COM displacement by the segment mass, and divided by the total body mass. Full body, posture, and leg COM onset was determined when velocity increased to above 2 SD or decreased below 2 SD determined from walkthrough trials.

Foot placement measurements were used to determine step time, step width, and step length. Step time was determined as the amount of time between heel contact of one foot to heel contact of the contralateral foot. Step length was determined as the displacement between heel contact of one foot to heel contact of the contralateral foot in the direction of the global x-axis. Step width was determined as the displacement between heel contact of one foot to heel contact of the contralateral foot in the direction of the global y-axis. In the case of a perturbed step, step lengths and widths were determined with respect to the final foot position following the perturbation. For statistical analysis, average walkthrough for each participant was subtracted from perturbation trials to determine changes in response.

Stability in the M/L direction was estimated by observing M/L COM location and velocity at heel contact. It has been stated that to determine stability, both COM position with respect to the base of support and velocity of the COM should be observed (Hof et al. 2005, You et al, 2001). COM-BOS in the M/L direction was measured from the lateral malleolus of the leading foot to the COM at heel contact. To determine the effect of M/L COM velocity on stability, the derivative of COM position was determined using the central difference equation. Average baseline measurements were subtracted from perturbation measurements to determine the effects due to the perturbation.

Muscle activity onset was determined as previously proposed in Marigold & Patla (2002) and Marigold et al. (2003). EMG onsets were determined only for initial perturbations for each condition. All raw electromyography signals were full-wave rectified and low pass filtered at 25 Hz using a single-pass 2^{nd} order

Butterworth filter. A perturbed muscle response profile for a surface translation trial was determined by subtracting the ensemble average profile of the control trials (obtained from the block of 10 control trials) from the slip trial. Muscle onset was determined as the time from perturbation onset to when the muscle activity increases above 2 SD for greater than 30 ms (to obtain an excitatory burst) or the time from perturbation onset to when the muscle activity decreases below 2 SD for more than 30 ms (to obtain an inhibitory burst). A muscle was required to be active 60% of trials to be considered a postural response due to perturbation as proposed in Henry, Fung, & Horak (1998a).

2.2.8 - Statistical Analysis

A direction (lateral, medial) by delay (heel contact, delayed) ANOVA was used for measurements of: perturbation onset, foot displacement during perturbation, stance time on the moving platform, ankle rotation during perturbation, COM differences between steps N and N+1 (full, posture, and leg), COM velocity onset (full, posture, leg), and lower limb joint angles. For measurements of step time, step length, step width, trunk pitch, trunk roll, COM position with respect to leading foot, and COM velocity with respect to leading foot, a direction (lateral, medial) by delay (heel contact, delayed) ANOVA was used for each step (N-1, N, N+1, N+2, N+3). To determine differences in COM position following the perturbation, a direction (lateral, medial) by delay (heel contact, delayed) ANOVA was used. In order to identify any sequencing of muscle onset during perturbations, separate repeated measures one way ANOVAs were used for each condition. All post-hoc analysis was examined using a Tukey test. Significance was determined at a level of 0.01. Outliers were identified and removed using studentized residuals and Cook's distance plots. These analyses will determine if any differences exist between lateral / medial perturbations and heel contact / delayed perturbations.

2.3.1 - Results

2.3.2 - Perturbation Characteristics

Perturbation onsets occurred at similar times in heel contact but not in delayed perturbations. Average perturbation onsets occurred at 0.17 +/- 0.03 sec, 0.32 +/- 0.06 sec, 0.16 +/- 0.03 sec, and 0.40 +/- 0.04 sec for lateral heel, lateral delayed, medial heel, and medial delayed perturbations respectively. ANOVA showed a significant interaction effect ($F_{(1,7)} = 25.18$, $p < 0.0001$) where significant differences occurred between timing of lateral delayed and medial delayed perturbations but not lateral heel and medial heel perturbations.

During perturbations, the perturbed foot was displaced less in delayed when compared to heel perturbations. A significant delay effect ($F_{(1,7)} = 39.80$, $p < 0.0001$) showed that the foot was perturbed less in delayed perturbations in both lateral and medial perturbations. Mean foot displacements during perturbation were 0.0764 +/- 0.0243 m and 0.0506 +/- 0.0181 m for heel and delayed perturbations respectively.

Stance times on the moving force plate were similar during perturbations. Mean stance time on the force plate was 0.0036 sec less during perturbation trials when compared to normal walking. ANOVA showed there were no significant differences in the stance times on the force plate between the perturbation conditions ($F_{(1,7)} = 2.74$, $p = 0.114$).

Ankle internal and external rotations occur during perturbations. A significant direction effect was observed in foot angle difference from heel contact on the force plate and the step following the perturbation ($F_{(1,7)} = 15.34$, $p < 0.0008$). Ankle difference was 23.3 +/- 44.5 degrees greater than walkthrough foot angle in lateral perturbations and 15.7 +/- 47.9 degrees less in medial perturbations.

2.3.3 - Step Characteristics

In the step following the perturbation, step time was shorter in the lateral surface translations when compared to medial and in heel contact perturbations when compared to delayed perturbations. No significant effects were seen in steps N-1 ($F_{(1,7)} = 0.46$, $p = 0.5058$), N ($F_{(1,7)} = 0.01$, $p = 0.9318$), N+2 ($F_{(1,7)} = 0.16$, $p = 0.6912$), N+3 ($F_{(1,6)} = 0.48$, $p = 0.4976$). In step N+1, a significant direction effect ($F_{(1,7)} = 36.17$, $p < 0.0001$) was shown where lateral perturbation step times were shorter than medial and a delay effect ($F_{(1,7)} = 8.11$, $p < 0.0096$) where heel contact perturbations has a shorter step time than delayed perturbations.

Steps following lateral and heel perturbations were shorter than normal baseline walking. No significant differences were observed in steps N-1 ($F_{(1,7)} = 0.33$, $p = 0.5704$), N ($F_{(1,7)} = 2.17$, $p = 0.1560$), or step N+2 ($F_{(1,7)} = 0.26$, $p = 0.6122$). Step N+1 displayed significant perturbation direction ($F_{(1,7)} = 14.40$, $p = 0.0011$) and delay ($F_{(1,7)} = 11.83$, $p = 0.0025$) effects where steps were shorter in lateral and heel perturbations. A significant delay effect was shown in step N+3 ($F_{(1,6)} = 12.11$, $p = 0.0027$) where perturbations at heel contact had longer steps than delayed perturbations. When step lengths were compared to zero, it was observed that lateral heel step N+1 was significantly shorter than baseline step length ($p < 0.0001$) and lateral heel ($p < 0.0001$) and medial heel ($p < 0.0005$) N+2 steps were significantly shorter than baseline.

Lateral perturbations caused a widening of the step following the surface translation while medial perturbations caused a narrowing of the following step. No significant differences were observed in steps N-1 ($F_{(1,7)} = 0.34$, $p = 0.5667$) and N ($F_{(1,7)} = 1.91$, $p = 0.1820$). A significant interaction effect was observed in step N+1 ($F_{(1,7)} = 22.27$, $p < 0.0001$) where lateral heel perturbations caused a widening of step length. Direction effects were observed in steps N+2 ($F_{(1,7)} =

38.31, $p < 0.0001$) and N+3 ($F_{(1,6)} = 58.47$, $p < 0.0001$) where lateral perturbations were narrower than medial perturbations.

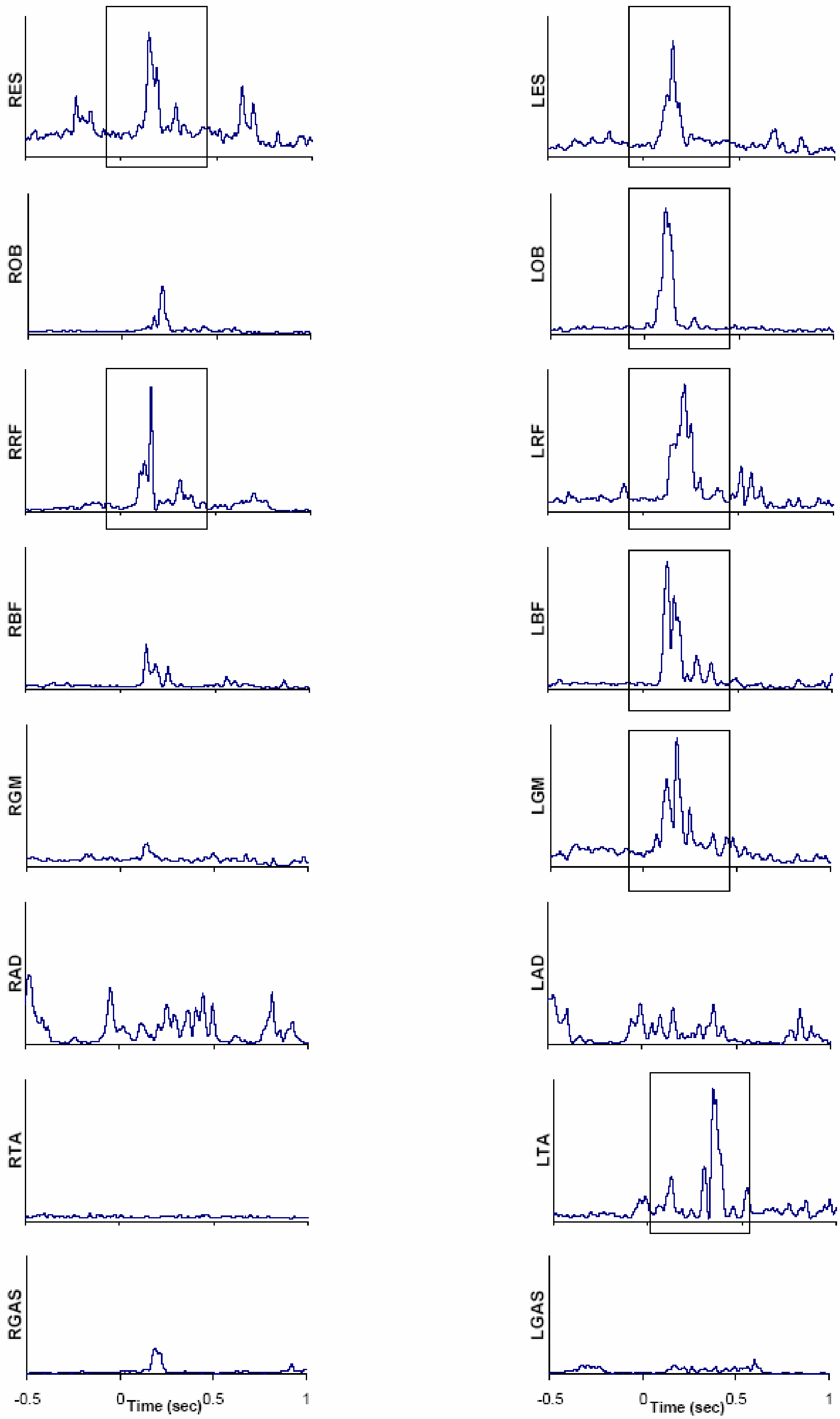
2.3.4 - Electromyography

Analysis of muscle onset suggests there is no order of muscle activation during frontal plane surface translations. Sample EMG profiles for lateral perturbations can be seen in Figure 2A and medial in 2B. Muscle EMG onset times can be viewed in Table 1. Analysis was only done on heel contact muscle onsets to determine any patterns of onset since two or less muscles were activated during the delayed perturbations. It was determined there were no significant differences in onset times for muscles during lateral heel contact perturbations ($F_{(8,48)} = 2.82$, $p = 0.1129$) or medial heel contact perturbations ($F_{(4,24)} = 1.34$, $p = 0.2983$). It can be observed that during lateral heel perturbations, most muscle activations are seen on the left side of the body while medial perturbations cause onsets mostly on the right side of the body. This can be seen in Figure 3.

Table 1: Onset times and SD for muscles following initial perturbations.

Condition	Muscle	Onset (sec)	SD
Lateral Heel	res	0.187	0.082
	rob	0.121	0.056
	rrf	0.123	0.022
	les	0.154	0.099
	lob	0.087	0.031
	lrf	0.109	0.019
	lbf	0.188	0.056
	lgm	0.165	0.034
	lta	0.228	0.168
Lateral Delayed	les	0.119	0.041
Medial Heel	res	0.254	0.121
	rob	0.206	0.128
	rrf	0.235	0.085
	rfb	0.211	0.070
	les	0.163	0.060
Medial Delayed	rob	0.163	0.118
	rrf	0.188	0.090

A



RES
ROB
RRF
RBF
RGM
RAD
RTA
RGAS
LES
LOB
LRF
LBF
LGM
LAD
LTA
LGAS
-0.5
0
Time (sec)
0.5
1

B

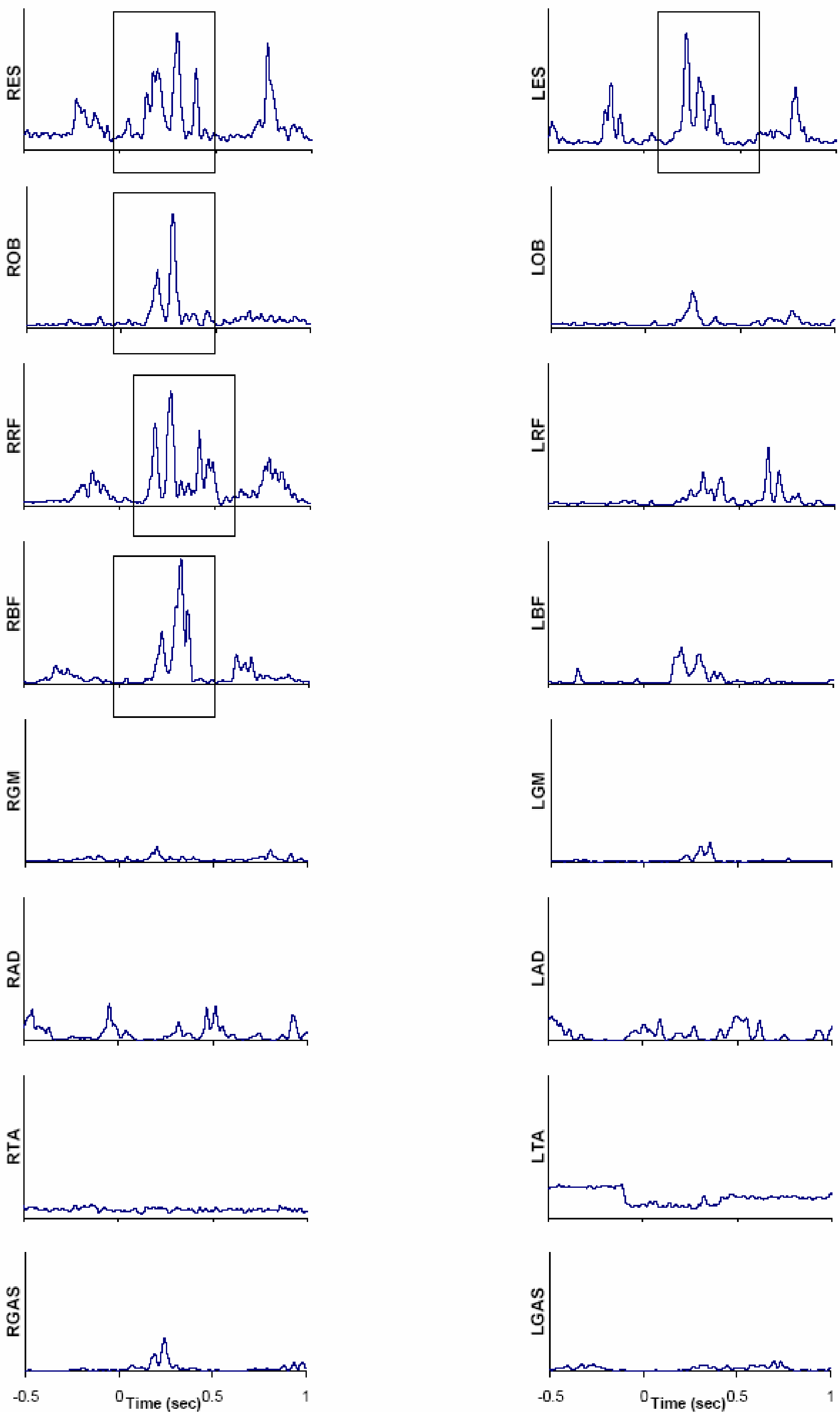

RES
ROB
RRF
RBF
RGM
RAD
RTA
RGAS
LES
LOB
LRF
LBF
LGM
LAD
LTA
LGAS
-0.5
0
Time (sec)
0.5
1
-0.5
0
Time (sec)
0.5
1

Figure 2: Sample full wave rectified and filtered (25 Hz) EMG profiles for lateral (in A) and medial (in B) heel contact perturbations. Muscles included are right erector spinae (res), left erector spinae (les), right oblique (rob), left oblique (lob), right rectus femoris (rrf), left rectus femoris (lrf), right biceps femoris (rbf), left biceps femoris (lbf), right gleutus medius (rgm), left gleutus medius (lgm), right adductor longus (rad), left adductor longus (lad), right tibialis anterior (rta), left tibialis anterior (lta), right gastrocnemius (rgas), and left gastrocnemius (lgas). Windows signify muscle responses.

A

B

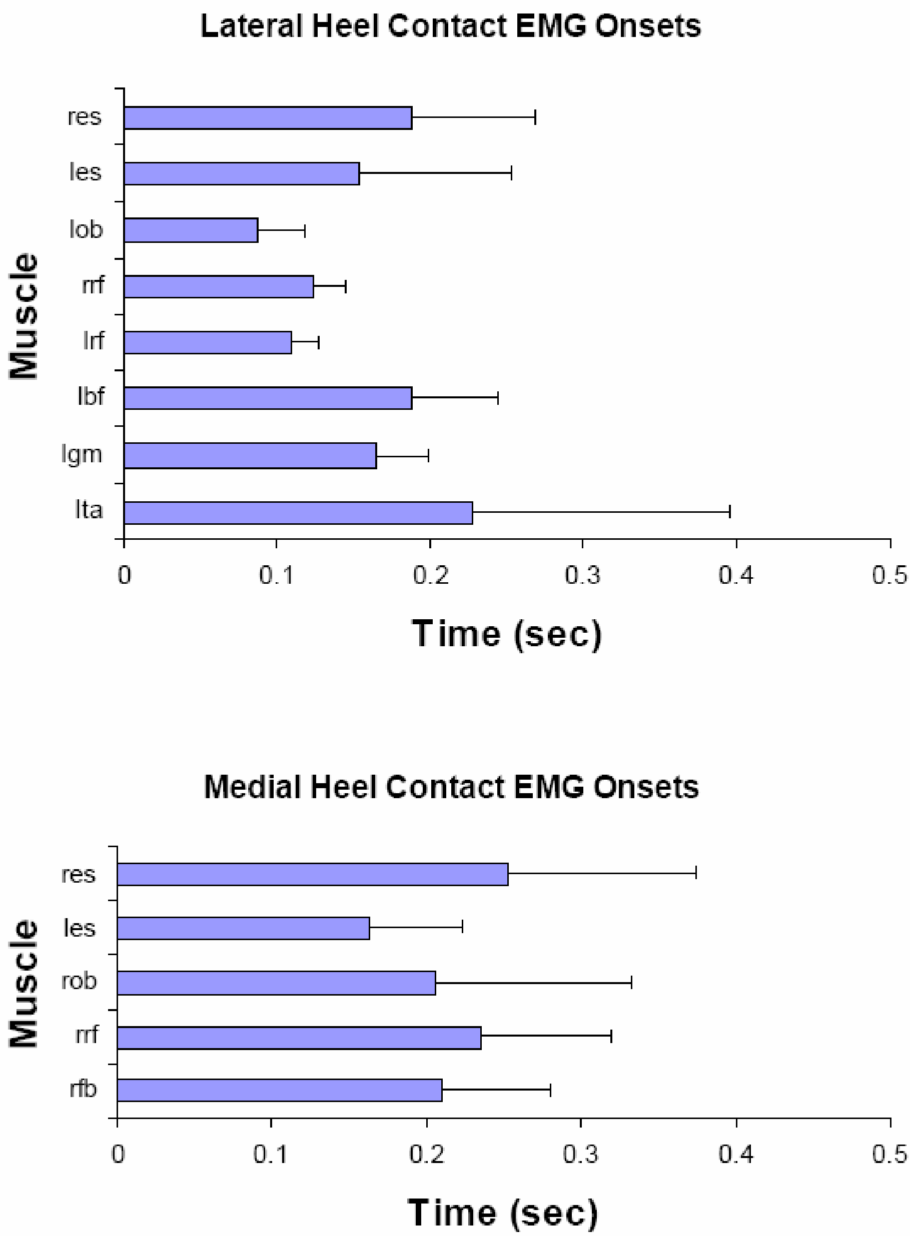


Figure 3: EMG onset times for initial perturbations. A: display of initial lateral heel contact perturbation. B: display of initial medial heel contact perturbation.

2.3.5 - Center of Mass Trajectories

Stick figure diagrams for all perturbations can be seen in Figure 4. Along with these stick figures, COM trajectories for each trial can be seen. These diagrams illustrate 1 step previous to perturbation and 2 steps following the perturbation.

Full body COM velocity onset times suggest full body COM was displaced later in delayed perturbations. Figure 5A illustrates how onset velocities were determined and Figure 5B displays onset times for full body, posture, and perturbed leg. Full body COM velocity onset showed a significant delay effect ($F_{(1,7)} = 11.00$, $p < 0.0033$) where onset occurred earlier in heel contact perturbations when compared to delayed perturbations. Posture COM velocity onset showed no significant effects ($F_{(1,7)} = 0.32$, $p = 0.5774$). Leg COM velocity onset showed an interaction effect ($F_{(1,7)} = 12.65$, $p < 0.0020$) where lateral delayed leg COM velocity onset occurred later than all other conditions. Onset times can be seen in Table 2.

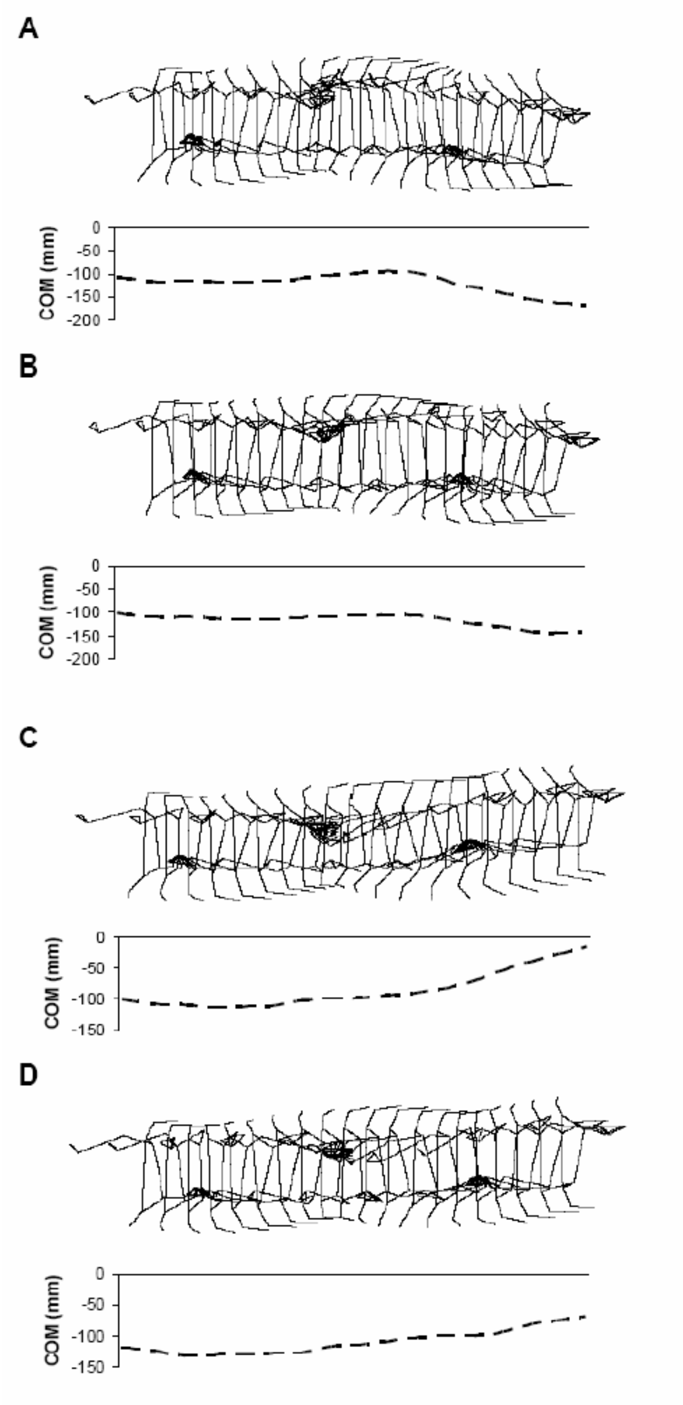


Figure 4: Top view stick figure of perturbations along with corresponding COM trajectories. A: lateral heel contact. B: lateral delayed. C: medial heel contact. D: medial delayed.

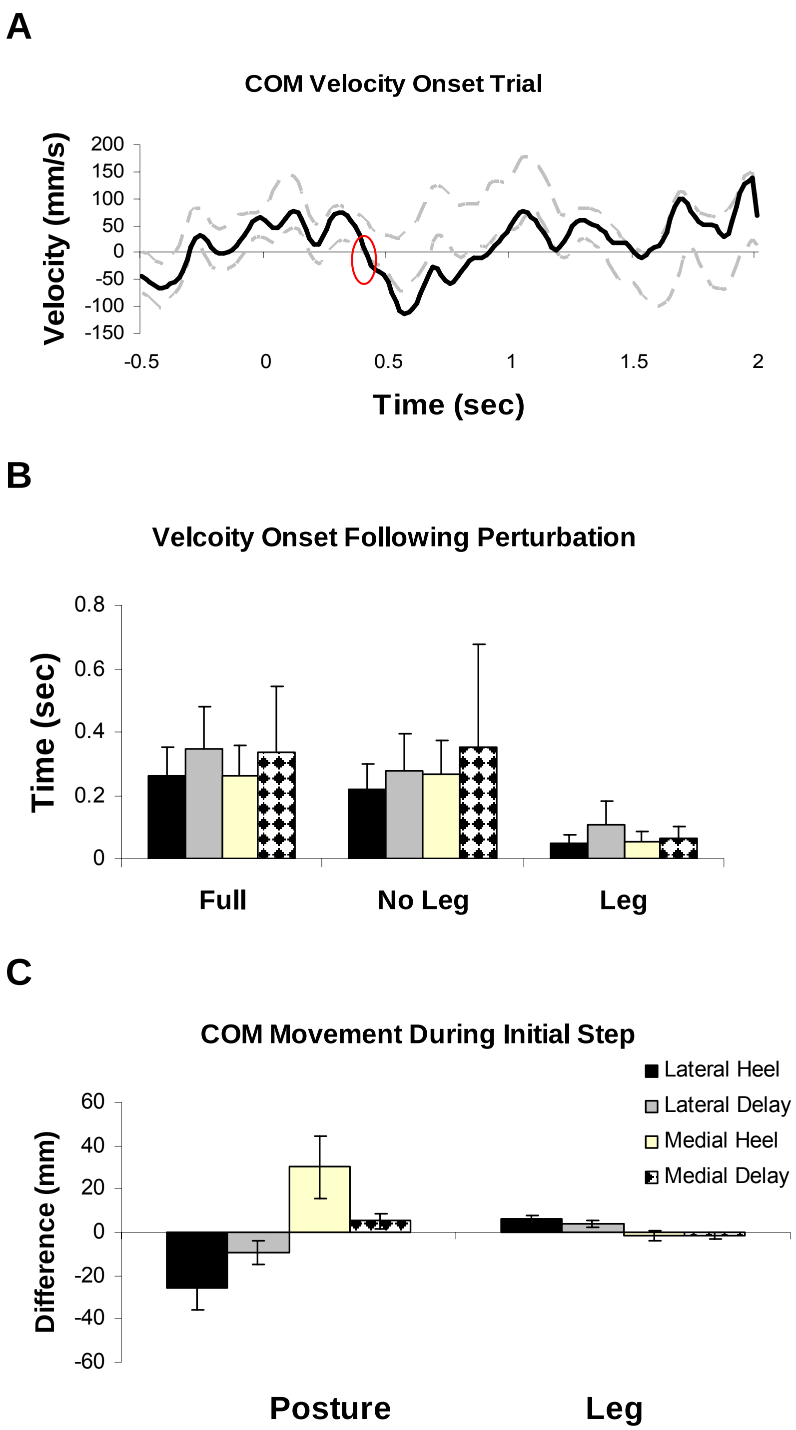


Figure 5: A: Display of how velocity onsets were determined. Solid line is velocity of full body COM along with +/- 2 standard deviations (dashed lines). The circle illustrates when the onset is determined. B: Plot of velocity onsets for full body COM, posture COM, and perturbed leg COM. C: Difference in posture and perturbed leg COM movement between step N and N+1.

Table 2: Velocity onset measures following perturbation.

Condition	Measure	Onset (sec)	SD
Lateral Heel	Full Body	0.261	0.089
	Posture	0.216	0.081
	Leg	0.049	0.025
Lateral Delayed	Full Body	0.346	0.132
	Posture	0.277	0.116
	Leg	0.109	0.070
Medial Heel	Full Body	0.264	0.093
	Posture	0.266	0.105
	Leg	0.054	0.032
Medial Delayed	Full Body	0.336	0.207
	Posture	0.352	0.327
	Leg	0.063	0.037

Differences in posture and leg COM between steps N and N+1 can be seen in Figure 5C. This was shown by significant direction effects in posture ($F_{(1,7)}$ = 88.48, $p < 0.0001$) and leg ($F_{(1,7)}$ = 102.91, $p < 0.0001$) displacements. To compare differences in magnitude, absolute values of leg and posture displacements were analyzed. ANOVA showed a significant delay effect in posture displacement ($F_{(1,7)}$ = 69.69, $p < 0.0001$) and leg displacement ($F_{(1,7)}$) = 10.47, $p < 0.0040$) where heel perturbations caused a greater posture displacement when compared to delayed. Leg displacement also showed a direction effect ($F_{(1,7)}$ = 41.50, $p < 0.001$) where lateral perturbations caused a greater leg displacement when compared to medial perturbations. To relate posture displacement to leg displacement, leg displacement was subtracted from posture displacement. Analysis displayed a significant delay effect ($F_{(1,7)}$ = 51.36, $p < 0.0001$) where displacement difference was greater in heel perturbations when compared to delayed perturbations.

An analysis of COM difference between perturbation onset and heel contact of step N+1 showed perturbed leg COM moved in the direction of the perturbation while COM of the posture would move in the opposite direction. Sample perturbed COM trajectories with walkthrough removed can be seen in Figure 6A. Full body COM trajectory was altered following surface translation perturbations. Figure 6B shows COM position at each heel contact following the perturbation. A repeated measures ANOVA showed heel perturbations had a greater deviation of COM when compared to delayed perturbations ($F_{(1,17)}$ =

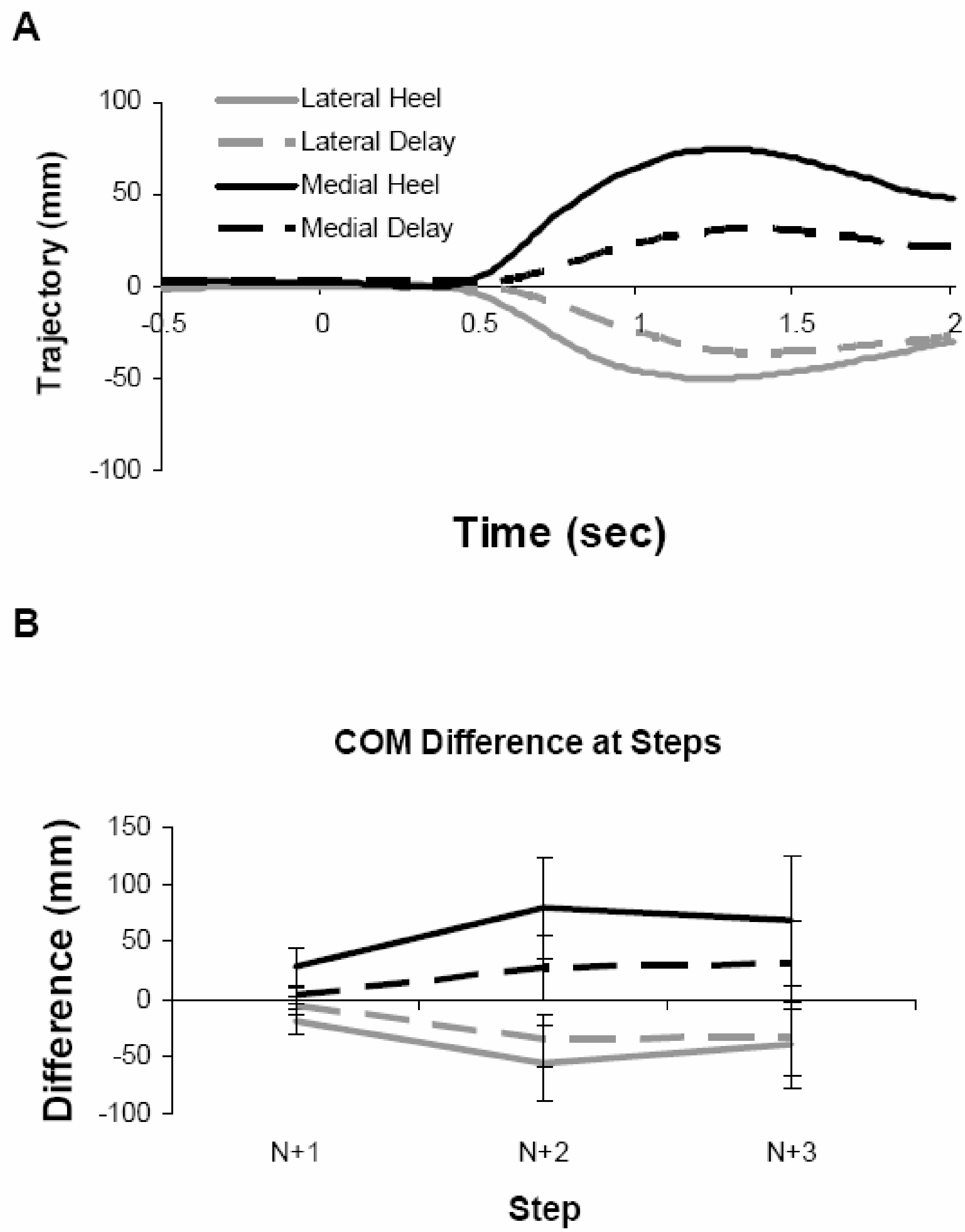


Figure 6: A: Sample M/L COM trajectories (walkthrough removed) for perturbations trials. B: Full body M/L COM location at each step following perturbations.

16.45, $p < 0.0008$). A significant step effect was also shown where significant differences occurred between steps N+1 – N+2 ($F_{(2,34)} = 110.59$, $p < 0.0001$) and N+1 – N+3 ($F_{(2,34)} = 54.27$, $p < 0.0001$) but no differences were seen between steps N+2 and N+3 ($F_{(2,34)} = 1.79$, $p = 0.1983$).

2.3.6 - Body Kinematics

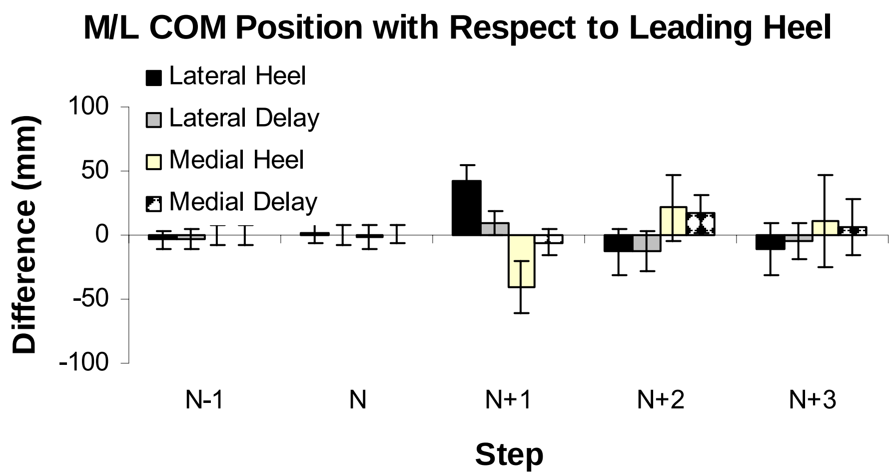
Trunk pitch was not altered following frontal plane surface translations. No significant differences were observed in steps N-1 ($F_{(1,7)} = 2.40$, $p = 0.1365$), N ($F_{(1,7)} = 0.98$, $p = 0.3331$), N+1 ($F_{(1,7)} = 1.91$, $p = 0.1810$), N+2 ($F_{(1,7)} = 0.02$, $p = 0.8902$), or N+3 ($F_{(1,5)} = 3.19$, 0.0944). Similarly, trunk roll did not change following frontal plane surface translations. No significant differences were observed in steps N-1 ($F_{(1,7)} = 0.43$, $p = 0.5174$), N ($F_{(1,7)} = 0.30$, $p = 0.5892$), N+1 ($F_{(1,7)} = 1.66$, $p = 0.2111$), N+2 ($F_{(1,7)} = 0.04$, $p = 0.8429$), or N+3 ($F_{(1,5)} = 0.25$, $p = 0.6229$).

Lower limb joint angles were altered between steps N and N+1 due to frontal plane surface translations. No significant changes were observed in right thigh flexion ($F_{(1,7)} = 0.30$, $p = 0.5882$), left thigh flexion ($F_{(1,7)} = 0.81$, $p = 0.3769$), or left ankle flexion ($F_{(1,6)} = 0.37$, $p = 0.5524$). Right thigh adduction showed an interaction effect ($F_{(1,7)} = 8.57$, $p < 0.0080$) where lateral perturbations had greater adduction than medial perturbations. Left thigh adduction showed an interaction effect ($F_{(1,7)} = 19.59$, $p < 0.0002$) where lateral heel perturbations had greater adduction than other perturbation conditions. Right knee flexion showed a direction effect ($F_{(1,7)} = 11.45$, $p < 0.0029$) where lateral perturbations had more knee extension than medial. Left knee flexion showed a delay effect ($F_{(1,7)} = 12.13$, $p < 0.0022$) where heel contact perturbations had greater knee extension than delayed. Right ankle flexion direction ($F_{(1,6)} = 15.50$, $p < 0.0010$) where lateral perturbations had greater dorsiflexion than medial perturbations and a delay effect ($F_{(1,6)} = 14.75$, $p < 0.0012$) where heel contact had greater dorsiflexion than delayed perturbations,.

2.3.7 - Dynamic Stability

In frontal plane surface translations, COM position with respect to the leading foot, was altered which would suggest changes in stability (Figure 7A). No significant differences were shown in steps N-1 ($F_{(1,7)} = 0.09$, $p = 0.7672$), N ($F_{(1,7)} = 2.55$, $p = 0.1249$), or N+3 ($F_{(1,6)} = 0.26$, $p = 0.6162$). A significant interaction effect was shown in step N+1 ($F_{(1,7)} = 30.71$, $p < 0.0001$) where no differences are shown between delayed perturbations while lateral heel perturbations had caused the COM to be further away from the leading foot and medial heel perturbations COM was closer to the leading foot. A direction effect was shown in step N+2 ($F_{(1,7)} = 56.33$, $p < 0.0001$) where lateral perturbations caused the COM to be closer to the leading foot when compared to medial perturbations.

A



B

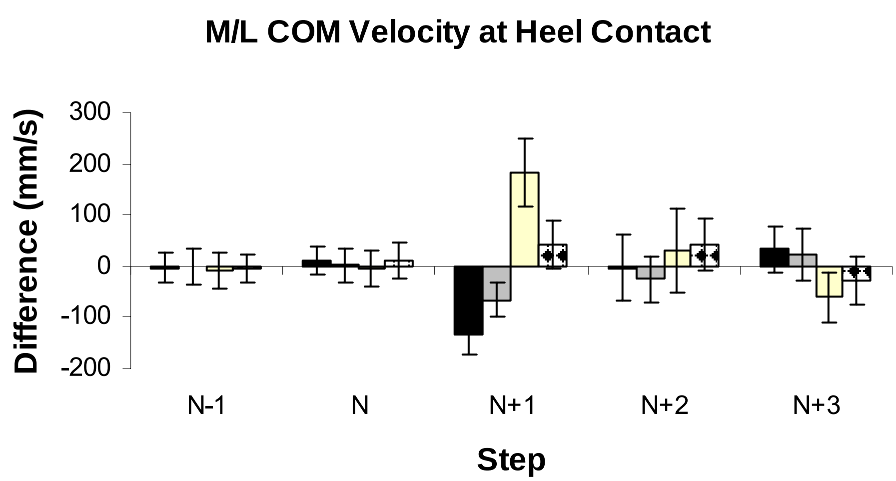


Figure 7: Determinants of M/L dynamic stability. A: Full body COM position with respect to leading leg. B: Full body COM velocity at each foot contact.

M/L full body COM velocity was altered at heel contact following the surface translation (Figure 7B). No significant differences were shown in step N-1 ($F_{(1,7)}$ = 0.01, $p < 0.9277$). Interaction effects were observed in step N ($F_{(1,7)}$ = 10.11, $p < 0.0047$) and step N+1 ($F_{(1,7)}$ = 29.77, $p < 0.0001$). In step N+1 all values were significantly different from each other except for lateral heel and lateral delay. Direction effects were observed in step N+2 ($F_{(1,7)}$ = 16.11, $p < 0.0006$) medial perturbations had a greater velocity than lateral and step N+3 ($F_{(1,6)}$ = 90.43, $p < 0.0001$) where lateral perturbations had a greater velocity than medial.

2.4.1 - Discussion

The present study examines frontal plane surface translations during locomotion. Previously, no studies have examined pure frontal surface translations nor have they studied perturbations occurring at differing times during the step cycle. This is an important element to examine due to the changes in stability which occur as the step cycle progresses. This study examined the effects of frontal plane surface translations to the right foot during locomotion. It was observed that there was some dissimilarity between the perturbation conditions. When stepping on the moving force plate, similar stance times were seen in each perturbation condition. This suggests the right foot was in contact with the ground for the same amount of time for each perturbation and no foot elevation responses are elicited by the CNS. Responses such as this are observed in tripping during early swing (Eng et al. 1994) and during cutaneous nerve stimulation of the foot in swing (Zehr et al. 1997). It would appear that the foot rides the platform during the perturbation. Similarly, participants would ride along a slip on a set of rollers on a walkway (Marigold and Patla, 2003). Since the foot was in contact for the same amount of time in each condition, it explains why the right foot was perturbed a less amount in the delayed perturbation conditions. Since the perturbation was elicited later in stance during the delayed conditions, the foot would “ride” the translating platform less, causing a decrease in perturbation distances. This decrease in perturbation distance may explain the decrease in effects on the measured parameters in the delayed conditions. One major consideration is the timing of the perturbation onset was slightly later in the medial delayed condition when compared to the lateral delayed perturbations. Even though the perturbation occurred at a later time, the foot was displaced the same distance in the delayed conditions. This consideration should be acknowledged when comparing onset times occurring after a perturbation.

2.4.2 - Muscle onsets show how stability begins to be corrected prior to stepping in heel contact perturbations

As with Marigold and Patla (2003), no distal to proximal sequencing was observed in the body. Observations from muscle activation show that during lateral heel contact perturbations, erector spinae and left oblique musculature are activated to stabilize the trunk. This stabilization is shown because no change occurs to trunk pitch and roll during the perturbation. The activation by the right rectus femoris causes extension in the right knee and may suggest a limb extension strategy during lateral heel contact perturbations. This is opposite to a flexion strategy seen in anterior slipping in Marigold and Patla (2002). The majority of the musculature which became active during lateral heel contact perturbations was in the left leg; the leg accepting the mass of the body following the perturbation. Since no direction effects were observed in the left knee, it can be suggested the CNS is using the musculature about the knee (rectus femoris and biceps femoris) to stabilize the knee during landing after the perturbation. This is different than what is seen in Marigold and Patla (2003) where an extensor strategy was seen in the unperturbed leg. This stabilization would ensure no fall would occur following the perturbation.

During medial heel perturbations, the majority of the muscle response is found in the right or perturbed limb. Again, the erector spinae and right oblique musculature is used to stabilize the trunk during the perturbation. Right rectus femoris and biceps femoris musculature activate during medial heel contact surface translations. Examining right knee kinematics shows that in medial perturbations, the right knee is less extended than in lateral perturbations. Therefore, the biceps femoris may have a higher activation to slightly flex the knee while the rectus femoris muscle uses co-contraction to also stabilize the knee. This co-activation seen in the stance limb may be used by the CNS to ensure the limb does not collapse during the perturbation.

No recognizable strategy was observed during delayed perturbations. This could occur because the CNS may not have appropriate time to make adjustments prior to contralateral heel contact since the perturbation occurs in late stance. In this study, delayed perturbations occurred at 0.32 and 0.40 sec for lateral and medial perturbations respectively. From this point, there would be approximately 0.1 sec remaining in the step. During this small amount of time, visual cues can not be used to make active changes. Changes that would occur during this time would be due to triggered responses due to proprioceptive information. This time would not be long enough to make significant changes to maintain stability and therefore, the CNS may primarily use steps following the perturbation to correct stability instead of triggered postural changes.

Muscle activations were only determined in the initial perturbations. This is because no sequencing could be determined in subsequent perturbations. Similar results were observed in Marigold and Patla (2003) when slipping occurred on a set of rollers. This decrease in muscular activity is evidence of predictive CNS control of muscle activations in perturbations following the initial responses to perturbation.

2.4.3 - CNS actively alters COM position in a direction opposite to perturbation

During perturbations at heel contact, the CNS uses reactive control to manipulate full body COM. The CNS compensates for perturbed leg movement by directing the COM in the opposite direction from the perturbation. Similar results were observed in Patla et al. (2002) when arm movement caused a perturbation on the body and Oddsson et al. (2004) during diagonal surface translations. Onset of posture COM velocity occurred between 216 and 352 ms. These onset times correspond to onsets of correction in instances of slipping observed by Cham et al. (2001). Evidence of reactive control by the CNS is seen by muscle activations occurring before changes in COM velocity. Muscle activations in this study appeared at 87 ms and continued until 254 ms during initial heel perturbations. Similar onset times were observed in Dietz et al. (1984) during locomotor corrections to rapid accelerations and decelerations of a treadmill. Latencies of this magnitude correspond mostly to polysynaptic or long latency reflexes (Pearson and Gordon, 2000). Oates et al. (2005) suggested that during gait termination on a set of rollers, latencies of this magnitude are from cutaneous receptors (Perry et al. 2000) and load-sensitive receptors of the foot (Misiasek et al. 2000). Only initial perturbations were examined because no identifiable responses could be determined in the perturbations following the initial. Similar results were observed in Nieuwenhuijzen et al. (2000), and Marigold and Patla (2002 and 2003). It has been suggested that this occurs due to a startle response employed by the CNS (Nieuwenhuijzen et al. 2000) or a possible overcompensation by the CNS during an initial perturbation which is followed by a fine tuning of muscular responses (Tang et al. 1998). In delayed perturbations, full body COM did not significantly deviate between perturbation onset and N+1 heel contact. This would suggest that there was no active control by the CNS due to the minimal time between perturbation and N+1 heel contact to make corrections.

2.4.4 - CNS actively controls ankle position during the surface translation

Many research studies have documented ankle dorsiflexion and plantarflexion during slipping (Cham and Redfern, 2001) and surface translations (Tang et al. 1998). Most analysis to date, has observed the body in a sagittal plane. Also, studies have assumed that perturbations are instantaneous and measurements during the perturbation have been ignored. Bothner et al. (2001) have suggested that perturbations cannot be assumed to be instantaneous. In this study, internal and external rotation of the ankle was observed during the course of the surface translation. As shown by the results, during lateral translations, the ankle was externally rotated and internally rotated during medial perturbations. These rotations of the ankle may be a display of the damping which is seen throughout the body decreasing the effects of the perturbation. For example, during a lateral perturbation an external rotation of the ankle would bring the heel closer to the midline of the body and in turn would decrease the effect of the perturbation on the leg. It is unlikely that this ankle rotation is passive because in a passive situation, the foot would travel without rotation on the force plate. This would suggest the CNS is actively attempting to lessen the effect of the perturbation on the body in an attempt to maximize stability.

2.4.5 - COM trajectory deviates following the surface translation

Following the perturbation in this study, full body COM continued to deviate in the direction opposite to the surface translation for two steps after the perturbation and a bias in travel path was displayed. A similar deviation can be seen in Figure 2A of Wall et al. (2002) during diagonal surface translations, but this bias was never addressed. This could occur due to the lack of visual cues present to straighten travel path or due to the inability for the CNS to regain steady state locomotion in three steps following the perturbation. During the experiment, there were no visual cues on the floor or a walkway for the participant to follow. This deviation could occur due to the lack of sensory information the CNS has to reorient walking trajectory. Another reason could be due to the inability for the CNS to regain stability. Results show that step width is constantly changing in order to correct for the perturbation. Since these corrections are being made for many steps following the perturbation, the CNS may be unable to maintain normal walking trajectory. A delay effect also shows that this bias in walking trajectory is decreased in delayed perturbations. In these perturbations, the leg is perturbed a lesser distance than in heel contact perturbations. This decrease in perturbation may have a lesser effect on the ability to regain normal locomotion and therefore less of an effect is seen on walking trajectory.

Even though the COM deviates following the perturbation, at no point does trunk pitch or roll deviate from normal walking trajectory. This is an interesting finding because it suggests the CNS puts trunk stability at a top priority during frontal plane perturbations. Trunk roll has been observed to reach a maximum deviation of 8 degrees within the first two steps following a diagonal surface translation (Oddsson et al. 2004). Differences seen in this study may occur because the velocity of the surface translation was much less than that used by Oddsson et al. (2004). In this study, the velocity of the surface was 0.3 m/s, while Oddsson et al. (2004) used velocities of 0.5 and 0.7 m/s. This may suggest that there is a velocity threshold where the CNS may be unable to maintain trunk angle during a surface translation.

2.4.6 - Continuous overcompensation of stability made by CNS

Observations from this study show that a surface translation in the frontal plane causes medio/ lateral instability for the step following the perturbation and up to three steps following. Pai et al. (1999) suggested that both COM-BOS position and velocity are determinants of balance in the saggital plane. You et al. (2001) also observed small COM-BOS distances during falls. In this study, this theory is being extended to the frontal plane. In lateral perturbations, the widening of step length causes full body M/L COM to be within the BOS, but this is accompanied with a large velocity towards the foot. Similar results were observed in Bhatt et al. (2005) during anterior surface translations and Oates et al. (2005) during gait termination on a set of rollers. This increase in M/L COM velocity could be very detrimental to stability because it could actually push the COM over the leading foot, possibly leading to a fall. This velocity difference was not significantly different between heel contact and delayed perturbations. In medial perturbations, the M/L COM is much closer to the leading foot and this is accompanied with a large velocity towards this foot, again possibly causing a major M/L instability. The steps following this initial step illustrate an overcompensation of the CNS in correcting stability. A similar overcompensation by the CNS can be seen in dynamic stability margin during locomotion on a compliant surface (MacLellan and Patla, 2006). M/L COM position in step N+2 was opposite to the initial response. Similar results are seen in M/L COM velocity and this overcompensation is observed 3 steps following the perturbation. This overcompensation suggests an inability for the CNS to accurately recover stability measures to baseline following a surface translation.

2.4.7 - Steady state locomotion is not achieved within 3 steps

Following perturbations during locomotion, previous evidence has shown that steady state can be regained following two compensatory steps. Results from this study indicate significant differences in step length, width, and COM control in 3 steps following the frontal plane translation. This difference may occur because a surface translation in the frontal plane may be more threatening than the previously observed perturbations. Similarly, Oddsson et al. (2004) observed that complete recovery from a diagonal surface translation occurs in five to six steps. When looking at step length and step width, it appears that the CNS is continually attempting to regain steady state locomotion following the perturbation. Step length is decreased following a lateral perturbation at heel contact but an increase is seen in step N+3. This would suggest an attempt by the CNS to regain control of locomotion. Similar results are seen in step width following the perturbation. A widening of step length in the step following the perturbation is seen in lateral perturbations and a narrowing is seen in steps N+2 and N+3. Similar results are observed in the medial direction perturbations. Due to the fact that not enough step measurements were collected, it is currently unknown how many steps are needed by the CNS to recover from a frontal plane surface translation.

2.4.8 - Reactive control is initially used to respond to perturbations followed by predictive control for muscle activities but not for kinematic variables

Previous research has indicated that the CNS initially uses reactive control to recover from an initial perturbation followed by feed-forward control to recover from subsequent perturbations (Marigold & Patla 2002; Oates et al. 2005; Pavol & Pai 2002; Pavol, Runtz, & Pai 2004). Adaptations that occur due to this feed-forward control include changes in COM position (Ferber et al. 2002, Marigold & Patla 2002; Pavol & Pai 2002; Pavol, Runtz, & Pai 2004) and EMG amplitude (Marigold & Patla 2002; Oates et al. 2005). In this study, increases in EMG amplitude were observed during the initial perturbation which is characteristic of the startle response (Nieuwenhuijzen et al. 2000). This would suggest the CNS uses predictive mechanisms to control following muscle activations as perturbations continue throughout the study. Observations of kinematic variables have shown that no differences occur as the study progressed. This may occur due to the CNS displaying a general response to recover from M/L surface translations in this study and outcomes from previous responses are not used in subsequent responses. In this study, two different directions and two different times of perturbation are elicited while in previous studies where adaptations occur, only one type of perturbation is used.

2.4.9 - Limitations

One major limitation in this study was the control of onset times of force plate movement during perturbation trials. When perturbations occurred, the movement onset of the force plate may not have been exactly at heel contact and at 200 ms following heel contact. Time was needed for the motor to generate enough power to initiate movement of the force plate. This limitation can be avoided by using a motor that would be able to generate enough power to move the force plate immediately. However, one was not available for this study. Another limitation associated with perturbation times was that a time of 200 ms had to be used to infer what the effects of the perturbation would be at push off. During pilot work, the average time of push off was found in ten baseline walkthrough trials and this time was used to elicit perturbations. Due to the problems with initiating movement immediately with the motor, this method would elicit perturbations after toe off on the moving force plate. This is why a delay of 200 ms was used.

Determining EMG onset times provided some limitations. After some observation, it could be seen that onset of muscle activity was dependent on the filtering frequency of the EMG signal. Studies looking at dynamic movements may cause changes to EMG signals and there is a debate of proper cutoff frequencies at this time. A frequency of 25 Hz was chosen because it has been used in previous work in order to be consistent with other studies in the laboratory. Along with a frequency of 25 Hz, frequencies of 10 Hz (which would resemble force generation in a muscle) and 100 Hz (which would resemble neural signaling to the muscle) were used in filtering and determining muscle onsets. These filter cut-offs yielded differing results for muscle activation following perturbation and could therefore change the conclusions obtained from their results. To correct this limitation, general guidelines for filtering EMG signals during locomotion studies should be determined.

Due to the spatial constraints in the laboratory, only three steps following the perturbation could be recorded. These spatial constraints did not allow an appropriate evaluation to determine the number of steps needed to return to a normal walking pattern following the perturbation. Use of a larger laboratory would allow the measurement of many steps following the perturbation and steady state locomotion could be determined in this case. Another limitation that occurred in the correction of locomotion following the perturbation was that no pathway was available. Since no pathway was available, participants had fewer visual cues to determine if they were walking in a straight line following the perturbation. During a real life slipping instance, people may be walking on a sidewalk or a pathway where visual cues are given and CNS responses may change in order to maintain stability and continue locomotion on the path provided. A study could be done with an outline of a path provided to determine if CNS responses change in order to maintain locomotion on the path.

2.4.10 - Future Directions

Conclusions from this study infer the mechanisms used by the CNS to maintain stability following a surface translation in the frontal plane. These conclusions include muscles used to regain stability, movement of the ankle during the perturbation, and foot placements following perturbations. From these observations, an exercise program could be developed in order to strengthen the muscles used recovering from these perturbations and increasing ankle range of motion in order to adapt to the perturbation. A future study could determine if falling is decreased after participation in a program that emphasizes improving the mechanisms needed to maintain stability following a frontal plane surface translation.

An interesting finding in this study was that observations suggested that feed-forward control of the CNS is not used in adapting to the perturbations in this study. It was suggested that this occurred because four differing perturbations were used in this study and the CNS was not able to adapt due to the variety of perturbations presented. This argument would be strengthened with further research examining a variety of perturbations given in an experimental session and determining if the CNS is able to adapt in these instances. Examples of projects can include perturbing the arm in differing direction during a reaching task or causing anterior or posterior perturbations during locomotion to determining any adaptation by the CNS.

Previously, it has been suggested that perturbations do not occur instantaneously and time is a component in surface translation studies. This study examined differences that occur during the course of the perturbation. Observations suggested that ankle rotation occurred during the perturbation in order to increase stability during the course of the perturbation. Observations such as this show how the CNS makes many immediate changes to posture during a frontal plane surface translation. It is very possible that this occurs in various types of perturbations. In the future, studies of perturbations should look more closely at changes made by the CNS during the course of the perturbation and how these immediate changes have an effect on the overall response used to regain stability following a perturbation.

2.5.1 - Conclusions

Surface translations can cause a great threat to stability due to the effects they have on the BOS. This thesis has examined how the CNS is able to regain stability following a surface translation in the frontal plane and a comparison was made between perturbations that occurred at heel contact and 200 ms following heel contact. A major finding was that in delayed surface translations, there was a decrease in muscle activity and body kinematics were affected much less when compared to heel contact perturbations. It is suggested that in delayed perturbations, the CNS does not have enough time to actively control stability prior to the step immediately after perturbation and stability is regained through placement of subsequent steps. With the limitations present, an analysis could not be performed to determine how many steps are needed to regain steady state locomotion. A second major finding was that it is possible the CNS was unable to use feed-forward mechanisms to adapt to the perturbation due to the variety of perturbations presented to the participant. Past research has only examined adaptations to one perturbation that occurs in randomly. Future studies could examine this point further to determine when feed-forward control cannot be used by the CNS.

3.1.1 – Adaptations of Walking Pattern on a Compliant Surface to Regulate Dynamic Stability

MacLellan, M.J. & Patla, A.E. (2006). Adaptations of walking pattern on a compliant surface to regulate dynamic stability. *Exp Brain Res*. 173(3): 521-30.

Adapting locomotor movements to the varied travel surface characteristics we encounter in our daily lives is essential. It is therefore not surprising that various researchers have examined changes in motor patterns while stepping on/off or traveling on compliant (Ferris et al. 1999; Dixon et al. 2000; Hardin et al. 2004; Moritz et al. 2004; Marigold and Patla 2005), uneven, or slippery travel surfaces. But the focus on what purpose these changes in motor patterns that invariably occur to serve is dependent on the researchers. For example, Hardin et al. (2004) focused on how metabolic cost during running on various travel surfaces is minimized by analyzing lower limb kinematic changes. Ferris et al. (1998) turned their attention to the control of vertical center of mass (COM) trajectory while running on a compliant surface and showed that leg muscle stiffness is regulated in order to maintain COM peak elevation. Dixon et al. (2000) argued that lower limb kinematic changes during running on a compliant surface serve to control peak impact force during heel contact.

How dynamic stability, the ability to maintain balance during locomotion, is controlled during travel on compliant surfaces has received much less attention. The importance of this issue is clearly highlighted by the demonstration that walking on compliant travel surfaces increases the risk of falls in the elderly (Lord and Menz 2000). Marigold and Patla (2005) examined the changes in lower limb trajectory that serve to minimize chances of tripping when stepping off a small area of compliant surface that affects a single step. How several steps on a compliant travel surface are regulated in order to maximize dynamic stability is not known. The central nervous system (CNS) deals with threats to stability through reactive, anticipatory, and predictive mechanisms. Dynamic stability is dependent on maintaining COM within a constantly changing and moving base of support (BOS) (Patla 2003). By examining stability during travel surface changes, it is possible to determine how and by which mechanisms the CNS maintains stability during these threats. Margin of stability is an important measure when determining stability because it relates COM and BOS directly. This measure has been extended by Hof et al. (2005) to include instantaneous COM velocity in the relation between COM and BOS. Theoretically, by using this analysis, the magnitude of stability threat can be determined by measuring how close the COM comes to the BOS. An unstable position is defined by the COM exceeding the BOS; the closer the velocity adjusted COM is to the BOS, the poorer the stability.

A compliant surface such as grass, sand, snow or a soggy field causes two perturbations to locomotion: an inability to use the kinesthetic system to accurately detect body orientation with respect to the travel surface and a mechanical perturbation which is caused by the compression of an extremely compliant viscoelastic surface during stepping. In this study, a foam mat is used as a walking surface for participants. The viscoelastic properties of the foam surface cause unpredictable reactions at the foot. It is unknown if the CNS corrects for surface unpredictability on a reactive step by step basis or plans ahead to compensate for unpredictability in order to maximize stability. This is the primary focus of this study. Towards that end we determine adaptations in step patterns, COM trajectory, and lower limb muscle activity when stepping onto and walking on a compliant surface. During locomotion on the compliant surface, it is hypothesized that the CNS will maximize stability by creating a larger BOS by manipulating step width and length to control COM within a larger area and by increasing toe elevation during swing phase on the travel surface that deforms.

3.2.1 - Methods

3.2.2 - Participants

Eight participants (5 females and 3 males; age 20.6±1.7 years; mass 66.2±15.2 kg) volunteered for this study. Participants had no muscular, neurological, or joint disorders which would affect their performance in this study. The study was approved by the Office of Research Ethics at the University of Waterloo and written informed consent was received by all participants.

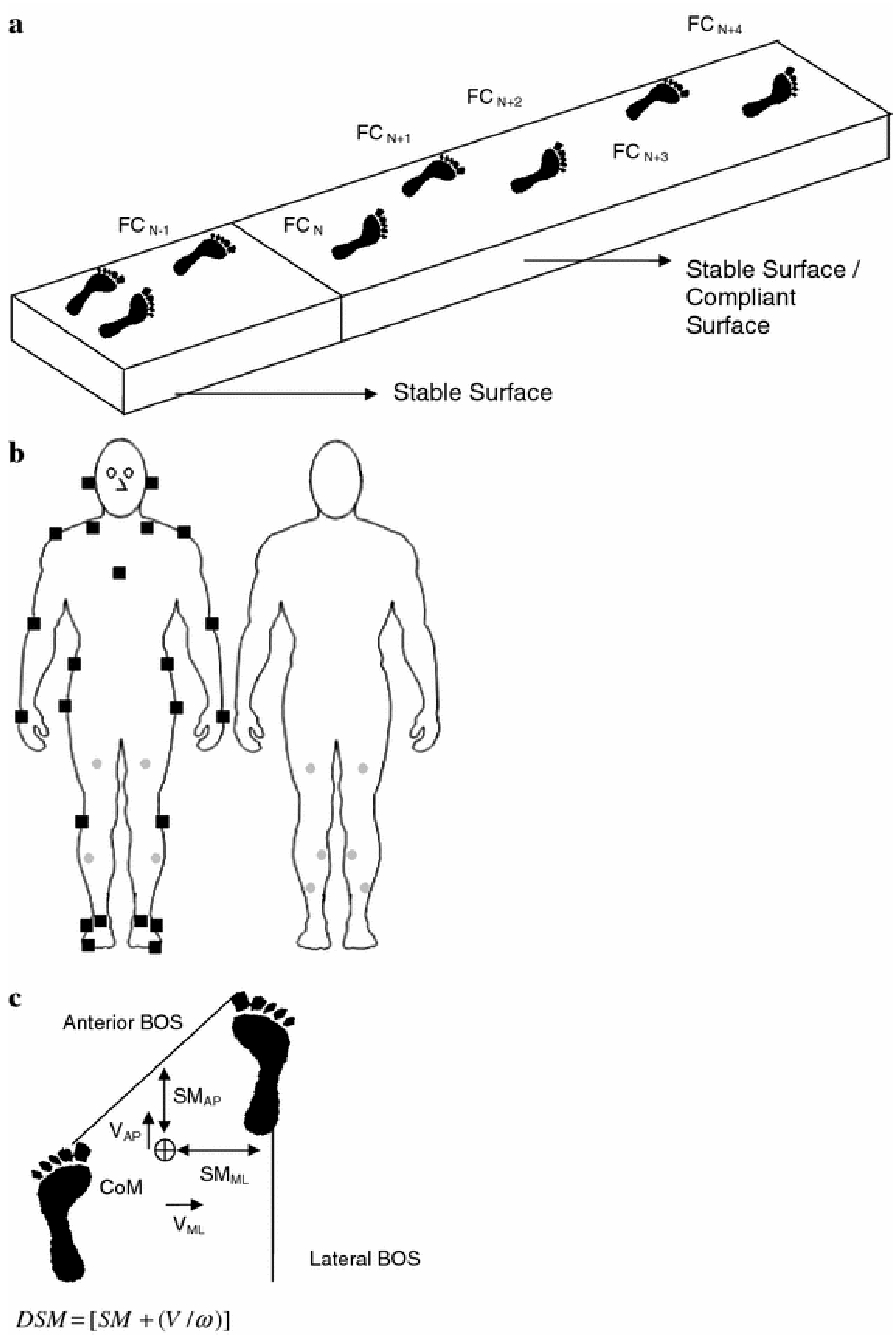


Fig. 1: Experimental setup used in this study. Part **a** identifies all foot falls and associates each step with a label. Part **b** illustrates iRED placements (*black squares*) and electromyography placements (*gray circles*) used in the study. Part **c** shows how stability margin (*SM*) is calculated in the anterior–posterior and medio-lateral directions

3.2.3 - Compliant surface

A medium density foam (5 m long, 0.91 m wide, 0.12 m depth) with a stiffness of 13.13 kN/m and a linear relationship between weight applied and surface compression (R-square value of 0.95) was used as the compliant walking surface in this study. An elevated wooden platform (1.1 m long, 1.2 m wide, 0.12 m depth) was used as a starting position during the compliant surface condition in order to avoid stepping up onto the surface. A diagram of the setup can be seen in Fig. 1a.

3.2.4 - Protocol

Two blocks of walking trials were collected: 10 baseline ground walking trials and 10 compliant surface walking trials. Baseline ground walking was completed first by all participants. During the baseline condition, thin lengths of green mat were placed on the ground to keep the walking area constant between the two conditions. Participants were instructed to start walking at a normal pace initiating gait with the left foot. Compliant surface trials were collected next.

Whole body kinematics were measured using three Optotrak cameras (Northern Digital Inc., Waterloo, Canada) at a frequency of 80 Hz. Twenty-three infrared emitting diodes (iREDs) were placed bilaterally on the following anatomical land marks: 5th metatarsal, superior anterior aspect of the foot, lateral malleolus, fibular head, greater trochanter, iliac crest, clavicle, acromion process, olecranon process, radial styloid process, and xiphoid process. iRED placements can be seen in Fig. 1b. This allowed us to construct a 12 segment model to estimate full body COM (Winter 2005).

Muscle activation was measured by surface electromyography (EMG) (Bortec, Calgary, Canada) from 10 lower limb muscles (rectus femoris, biceps femoris, tibialis anterior, medial gastrocnemius, and soleus bilaterally) using Ag–AgCl (Kendall Medi-Trace, Chicopee, MA, USA) electrodes. Signals were band pass filtered at 10–1,000 Hz and sampled at 1,200 Hz. The common mode rejection of the EMG system was 115 dB with an input impedance of 10 GΩ. Placement of iREDS and EMG electrodes can be seen in Fig. 1b. Muscle activation was only collected from seven of eight participants due to an equipment problem.

3.2.5 - Data analysis

All kinematic data from iRED markers were low-pass filtered at 7 Hz (dual-pass, second-order, Butterworth filter). Ankle markers were used to determine step width, step length, and step time. Steps were termed in relation to participant's initial step on the compliant surface. The initial step that participants took with their left foot was termed FC $_{N-1}$, and the following step with the right foot (the first step on the compliant surface) was termed FC $_N$ with FC $_{N+1}$, FC $_{N+2}$, FC $_{N+3}$, and FC $_{N+4}$ following. Percent variability for step width and length was determined by dividing the standard deviation by the mean measure for each step. Step velocity was calculated by dividing step length by step time during each step. Toe trajectory was determined for each participant using the iRED placed on the 5th metatarsal. From these trajectories, initial maximum and minimum values were determined for the swing phase of each step. Distance from the lateral malleolus to vertical COM position was calculated at each heel contact. A 12-linked segment model (head, trunk, upper arm, lower arm and hand, thigh, shank, and feet) was used to estimate COM in the vertical, medio-lateral, and anterior–posterior directions (Winter 2005). Vertical and medio-lateral COM trajectories were normalized to percent of left stride and position of COM was relative to the initial step taken on the compliant surface. This created two normalized bins for COM data (bin 1 consisted of data from FC $_{N-1}$ to FC $_{N+1}$, termed stride 1, bin 2 from FC $_{N+1}$ to FC $_{N+3}$, termed stride 2). An analysis of trajectory slopes was used to determine magnitudes and times of minimum and maximum peaks in the vertical and medio-lateral directions. Using three trunk markers (right clavicle, left clavicle, and xiphoid) three-dimensional trunk angles were determined at each step using the Cardan method of determining angles. Dynamic stability margin was calculated as described by Hof et al. (2005) (Fig. 1c). Dynamic stability margin in the anterior–posterior direction (DSM_{AP}) was measured as the distance from the line joining the toes (anterior border of the BOS) to the COM at heel contact added to the instantaneous anterior–posterior velocity of the COM divided by the square root of the height to the COM divided by gravity. Stability margin in the medio-lateral direction (DSM_{ML}) was measured from the lateral malleolus of the leading foot (lateral border of the BOS) to the COM at heel contact added to the instantaneous medial-lateral velocity of the COM divided by the square root of the height to the COM divided by gravity. This measurement was calculated for each heel contact.

EMG data were full-wave rectified and low-pass filtered at 10 Hz (dual-pass, second-order, Butterworth filter), normalized, and binned. These bins of EMG were broken down into functional bins (0–30%: weight acceptance, 30–50%: push-off, 50–80%: early swing, 80–100%: late swing). Average EMG (AEMG) was calculated for each of these functional bins for each muscle.

3.2.6 - Statistical analysis

Repeated measures analysis of variance (ANOVA) was used on all data sets. A condition (ground, foam) by step (FC_{N-1}, FC_{N}, FC_{N+1}, FC_{N+2}, FC_{N+3}, FC_{N+4}) ANOVA was used to determine differences in step characteristics, three-dimensional trunk angles, maximum toe trajectory, minimum toe trajectory, and stability margin data. For COM data, condition (ground, foam) by peak (max_1, min_1, max_2, min_2, max_3, min_3, max_4, min_4) ANOVA was used to determine the differences for vertical COM displacement and a condition (ground, foam) by peak elevation (left$_1$, right$_1$, left$_2$, right$_2$, left$_3$) ANOVA was used for medio-lateral COM displacement. Analyses for EMG data were performed for each muscle and each stride. A condition (ground, foam) by functional bin (weight acceptance stride 1, push-off stride 1, early swing stride 1, late swing stride 1, weight acceptance stride 2, push-off stride 2, early swing stride 2, late swing stride 2) ANOVA was used to determine any significant differences. ANOVA was also administered for measurements of variance. To determine trial effects for all variables, measurements from the initial and final trial were analyzed in a condition (ground, foam) by trial (first, last) by step (FC_{N-1}, FC_N , FC_{N+1}, FC_{N+2}, FC_{N+3}, FC_{N+4}) ANOVA. For all analyses, an alpha level of 0.05 was used to determine significance. Post hoc analysis was conducted when significant differences existed using a least squares difference test with an alpha level of 0.05.

3.3.1 - Results

3.3.2 - Whole body center of mass

Vertical whole body COM follows a different trajectory on the compliant surface when compared to stable ground. Stick figure trajectories in Fig. 2 display representative trials. Estimated whole body COM trajectories can be seen in Fig. 3. An analysis of vertical COM trajectory peaks revealed an interaction

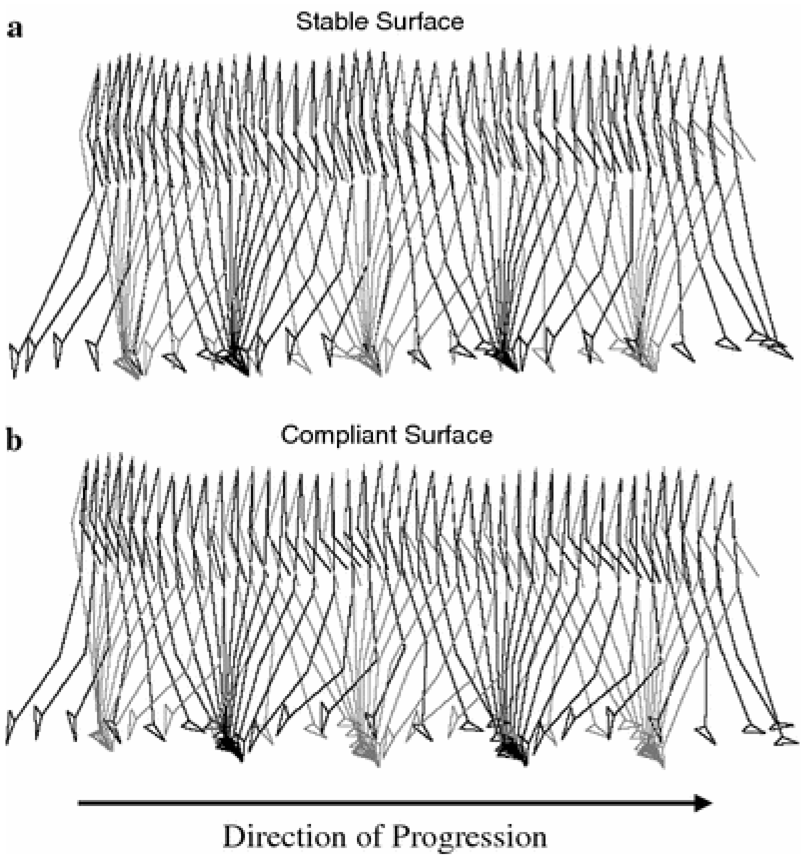


Fig. 2: Representative trial stick figure diagrams illustrating movement on stable ground (**a**) and on the compliant surface (**b**). Position of foot markers are 5th metatarsal, lateral malleolus, and superior anterior aspect of the foot

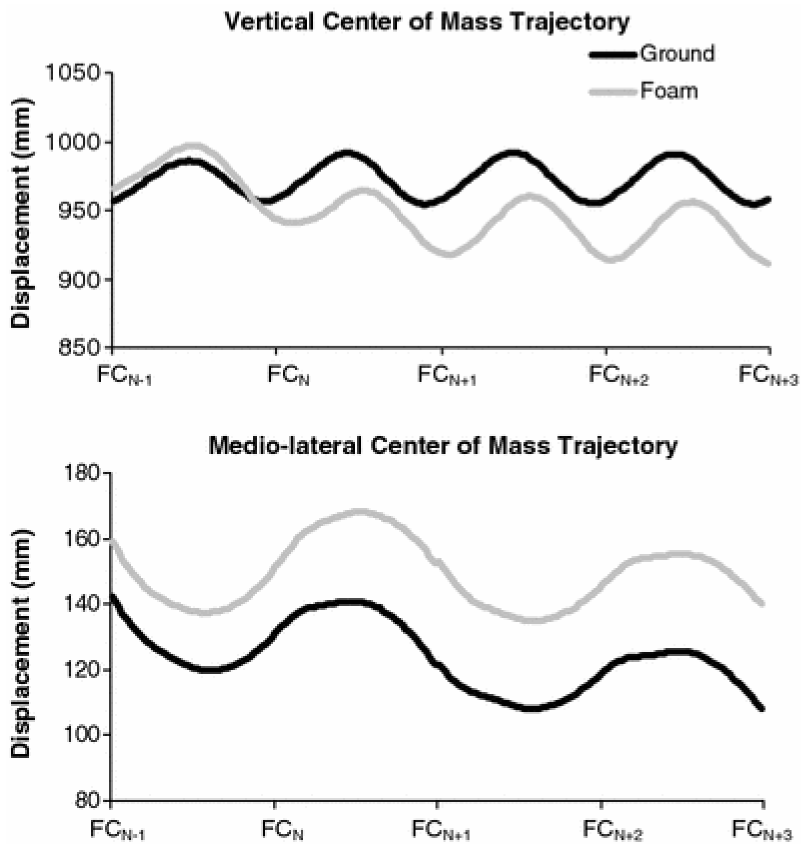


Fig. 3: Average center of mass trajectories for ground (*black*) and compliant surface (*gray*) conditions. Note that at foot contact *N*-1, the heel coordinate was set at 0,0: this allowed us to compare changes in subsequent COM trajectory for the two travel surfaces

effect in peak magnitude ($F_{(7,49)}$=24.87, P<0.0001). Post hoc analysis inferred that the initial maximum was significantly greater on the compliant surface and the remaining peaks were significantly lower. To determine if the lowering of COM was primarily due to compression of the compliant surface or due to active coordination of joint angles lowering the center of mass, an analysis was done to determine the vertical position of the COM with respect to the ankle marker. Mean values of ankle marker and COM difference were 0.865 and 0.859 m for ground and compliant surfaces, respectively. The analysis showed that COM is significantly lower in the compliant surface condition ($F_{(1,7)}$=6.25, P<0.041) through a condition effect. This illustrates that the decrease seen in vertical COM is not just due to the compression of the travel surface, but involves active lowering of the COM. Analysis of peak time displayed a significant interaction effect ($F_{(7,49)}$=15.55, P<0.0001). This difference caused a phase lag in the compliant surface COM trajectory with respect to the ground trajectory. No significant differences were seen in trajectory peak magnitude or peak time in the medio-lateral direction. A trial effect was produced when examining whole body vertical COM peaks. Whole body vertical COM tended to decrease as trials progressed while walking on the compliant surface which was seen in a condition effect ($F_{(1,5)}$=50.09, P<0.0009). No significant trial effects were observed in medio-lateral COM trajectory, vertical COM peak time, or medio-lateral COM peak time.

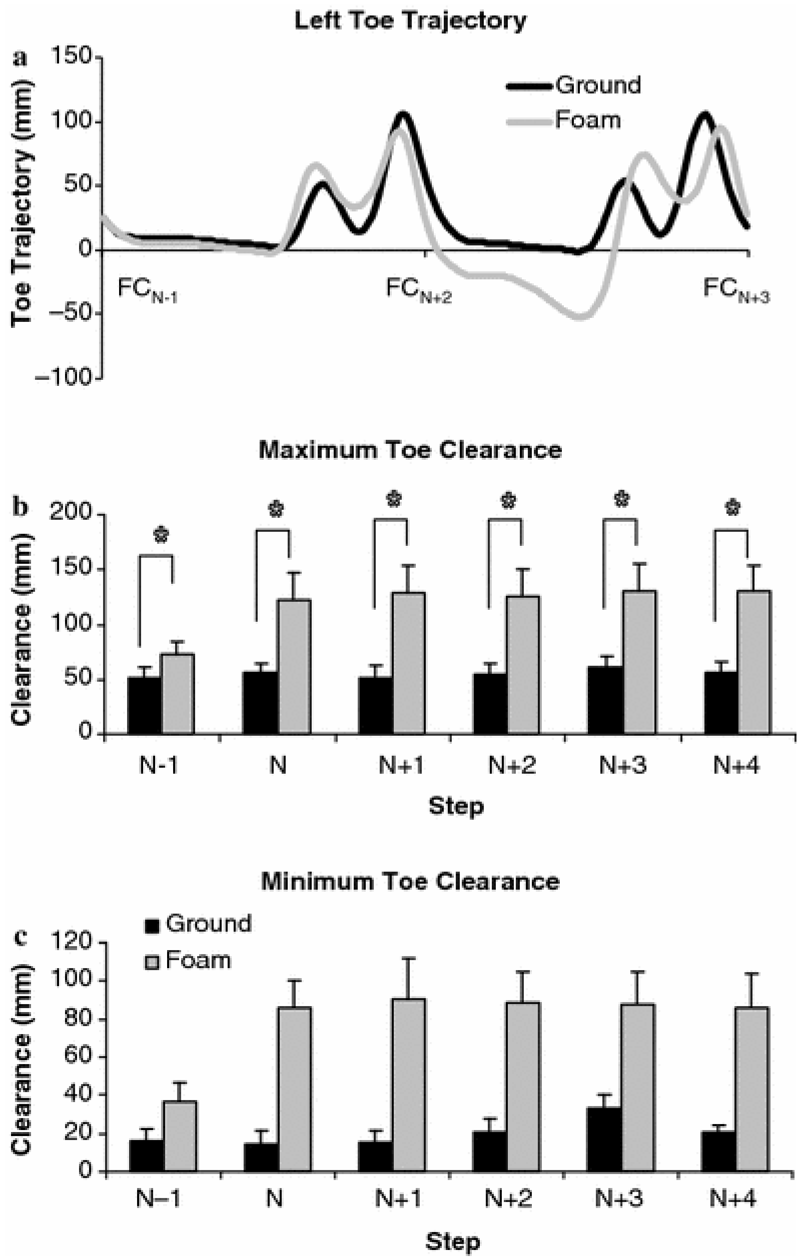


Fig. 4: Part **a** illustrates representative trial toe trajectories (determined from left 5th metatarsal marker) for ground (*black*) and compliant surface (*gray*) conditions. Parts **b** and **c** show average maximum and minimum toe clearance (determined from left 5th metatarsal marker) in the ground (*black*) and compliant surface (*gray*) conditions. Significant differences are denoted by *

3.3.3 - Toe trajectory

Initial maximum and minimum peak toe trajectories tended to increase during the compliant surface condition. Representative toe trajectories can be seen in Fig. 4a. Initial maximum peak values during each step are illustrated in Fig. 4b. These peaks were significantly larger in each step on the compliant surface which was seen in a condition effect ($F_{(5,35)}$=39.61, P<0.0001). No significant differences were displayed between trials. Minimum trajectory peak values are shown in Fig. 4c and these peaks are significantly larger in the compliant surface condition ($F_{(5,35)}$=49.06, P<0.0001). No significant differences were displayed between trials.

3.3.4 - Step characteristics

Step widths, lengths, and times tended to increase while walking on the compliant surface. Table 1 displays step width, length, times, and variances for each measure. An interaction effect was observed for step width ($F_{(4,28)}$=3.59, P<0.018) and step length ($F_{(4,28)}$=3.41, P<0.022). Post hoc analysis confirmed that all step widths and lengths were significantly larger than their corresponding ground steps. Interesting to note is that width for FC_N on the compliant surface was significantly larger than FC_{N+1}, FC_{N+2}, and FC_{N+3}. An interaction effect ($F_{(4,28)}$=5.07, P<0.004) illustrated that the variance for compliant surface FC_{N+2} was larger than FC_N , FC_{N+1}, and FC_{N+4}. Examination of step length variance yielded a condition effect ($F_{(1,7)}$=5.72, P<0.049) where step length variance was significantly larger on the compliant surface. Step time increases while walking on a compliant surface ($F_{(4,28)}$=4.78, P<0.0047) which was seen in astep by surface interaction effect. Post hoc analysis revealed that the compliant surface FC $_{N+4}$ time was significantly smaller than FC $_{N}$, FC $_{N+1}$, and FC $_{N+2}$. Step time variance showed an interaction effect where step time variance was larger in compliant surface FC $_{N+1}$, FC $_{N+2}$, FC $_{N+3}$, and FC $_{N+4}$ when compared to corresponding ground steps ($F_{(4,28)}$=4.35, P<0.0074). No trial effects were seen in step width, length, or time. An interaction effect was seen in step velocity ($F_{(4,28)}$=4.37, P<0.0072) in which step velocity was greater in the compliant condition in FC $_{N}$ and FC $_{N+4}$ and greater in the ground condition in FC $_{N+3}$.

Table 1: Average measurements for step width, length, time, and variability for each step in ground and compliant surface conditions

Step	Variable	Ground	Foam	*P* value
N	Step width (mm)	230.756	262.617	< 0.0001
	Step width variance (SD^2)	304.585	430.503	0.0892
	Step length (mm)	653.494	682.833	0.0001
	Step length variance (SD^2)	697.125	1,130.532	0.0694
	Step time (s)	0.603	0.586	0.526
	Step time variance (SD^2)	0.009	0.012	0.3219
N+1	Step width (mm)	239.938	251.306	0.0386
	Step width variance (SD^2)	438.647	542.837	0.173
	Step length (mm)	700.981	760.619	< 0.0001
	Step length variance (SD^2)	611.322	1,075.026	0.0923
	Step time (s)	0.566	0.612	0.0001
	Step time variance (SD^2)	0.007	0.009	0.0113
N+2	Step width (mm)	222.404	234.098	0.0338
	Step width variance (SD^2)	319.747	1,015.549	< 0.0001
	Step length (mm)	692.589	744.931	< 0.0001
	Step length variance (SD^2)	431.570	1,070.151	0.018
	Step time (s)	0.562	0.592	0.0064
	Step time variance (SD^2)	0.007	0.013	0.0001
N+3	Step width (mm)	235.254	247.260	0.0296
	Step width variance (SD^2)	354.030	762.478	< 0.0001
	Step length (mm)	704.414	751.857	< 0.0001
	Step length variance (SD^2)	339.274	1,135.012	0.0007
	Step time (s)	0.544	0.594	0.0004
	Step time variance (SD^2)	0.007	0.019	< 0.0001
N+4	Step width (mm)	228.377	255.886	< 0.0001
	Step width variance (SD^2)	425.116	625.392	0.0822
	Step length (mm)	692.819	730.583	< 0.0001
	Step length variance (SD^2)	428.158	887.752	0.0069
	Step time (s)	0.560	0.568	0.4614
	Step time variance (SD^2)	0.007	0.012	< 0.0001

3.3.5 - Lower limb electromyography

EMG was analyzed for each muscle and for each side independently. Figures 5 and 6 display significant muscle activities in the left and right leg, respectively. It was observed that rectus femoris had no significant differences in the left leg although an interaction effect was seen in the right side ($F_{(7,42)}$=2.84, P=0.0162). In the right side, increases in AEMG were seen during weight acceptance in both strides in the compliant condition. Biceps femoris displayed an interaction effect for the left side ($F_{(7,42)}$=5.61, P<0.0001) and for the right side ($F_{(7,42)}$=3.52, P<0.0046). Post hoc analysis illustrated that biceps femoris activity was significantly greater during left swing in stride 1 and stance and late swing in stride 2 in the compliant surface condition. In the right side, bicep femoris was significantly greater in the compliant surface condition during late swing in stride 1 and weight acceptance and late swing in stride 2. Tibialis anterior muscle activation also had interaction effects on the left side ($F_{(7,42)}$=5.2, P<0.0003) and the right side ($F_{(7,42)}$=6.23, P<0.0001). Post hoc analysis showed that tibialis anterior muscle activation was significantly greater on stable ground during late swing in stride 1 and stride 2. In the right leg, tibialis anterior activity was greater on the compliant surface in late swing in stride 1 and stride 2. Right tibialis anterior activity was greater in the compliant surface condition during stance in stride 1 and late swing in stride 2. Gastrocnemius yielded interaction effects for the left side ($F_{(7,42)}$=3.43, P<0.0054) and right side ($F_{(7,42)}$=10.07, P<0.0001). Left gastrocnemius activity was greater in the ground condition during stance in

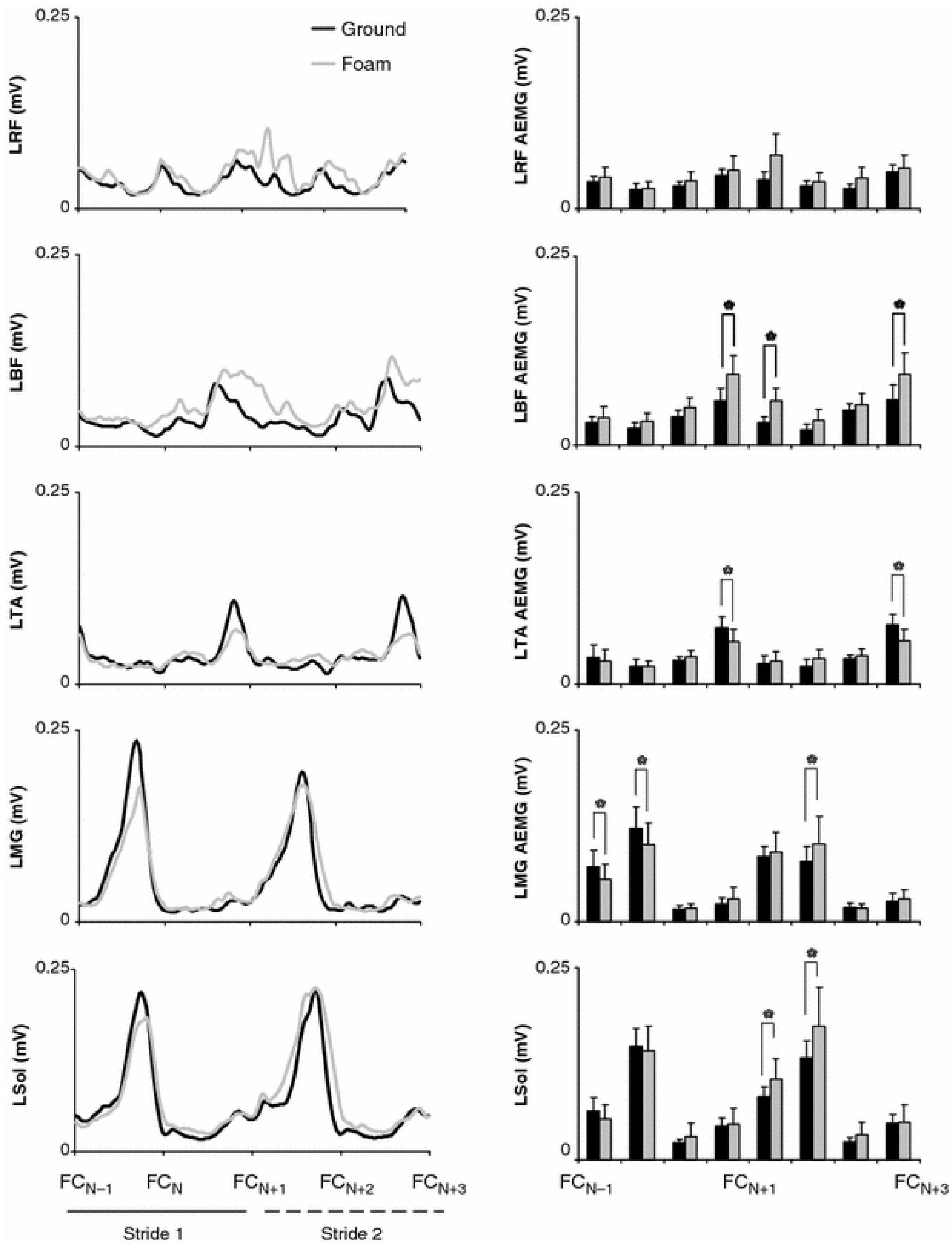


Fig. 5: Full-wave rectified and filtered (*left*) and averaged (*right*) electromyography for the left leg. Average electromyography (*AMEG*) is calculated for the functional bins stated in Methods (weight acceptance stride 1, push-off stride 1, early swing stride 1, late swing stride 1, weight acceptance stride 2, push-off stride 2, early swing stride 2, late swing stride 2). Muscle activities are illustrated for left rectus femoris (*LRF*), left biceps femoris (*LBF*), left tibialis anterior (*LTA*), left medial gastrocnemius (*LMG*), and left soleus (*LSol*). Significant differences are denoted by *

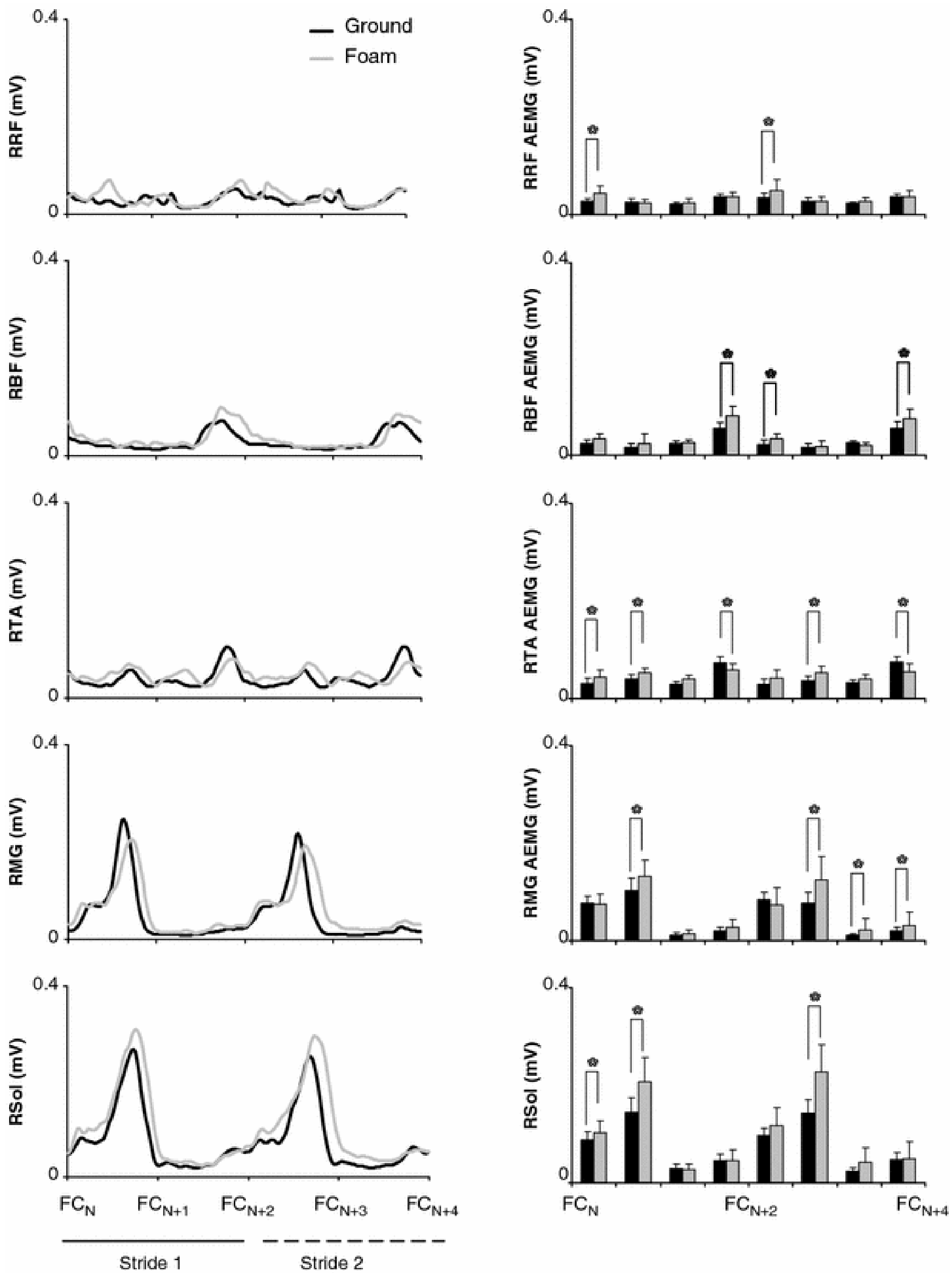


Fig. 6: Full-wave rectified and filtered (*left*) and averaged (*right*) electromyography for the right leg. Average electromyography (*AMEG*) is calculated for the functional bins stated in Methods (weight acceptance stride 1, push-off stride 1, early swing stride 1, late swing stride 1, weight acceptance stride 2, push-off stride 2, early swing stride 2, late swing stride 2). Muscle activities are illustrated for right rectus femoris (*RRF*), right biceps femoris (*RBF*), right tibialis anterior (*RTA*), right medial gastrocnemius (*RMG*), and right soleus (*RSol*). Significant differences are denoted by *

stride 1 and greater in the compliant surface in late swing in stride 2. Right gastrocnemius activity was greater in the compliant surface condition during late stance in stride 1 and late stance and swing during stride 2. Soleus EMG displayed interaction effects for the left side ($F_{(7,42)}$=4.74, P<0.0005) and right side ($F_{(7,42)}$=7.41, P<0.0001). Analysis showed that left soleus activity was greater in the compliant condition during stance in stride 2. In the right soleus, activity was greater in stance during stride 1 and late stance during stride 2 in the compliant condition.

3.3.6 - Stability margin

Dynamic stability margin in the anterior–posterior direction (DSM_{AP}) was significantly different in the compliant surface condition which was shown by a condition by step interaction effect (F(4,28)=10.91, P < 0.0001). Figure 7a illustrates measurements of DSM_{AP}. Post hoc analysis revealed that DSM_{AP} was significantly greater during FC_{N} , FC_{N+2}, and FC_{N+4}. DSM_{AP} variance was greater on the compliant surface which was seen in a condition effect ($F_{(4,28)}$=15.30, P<0.0058). No differences were seen in DSM_{ML} between the compliant surface and ground condition. Variability in DSM_{ML} was much different in the two conditions which was revealed in a condition by step interaction effect ($F_{(4,28)}$=3.65, P<0.0163). Post hoc analysis revealed that DSM_{ML} variability was significantly greater in compliant condition FC_{N} , FC_{N+2}, FC_{N+3}, and FC_{N+4}, with

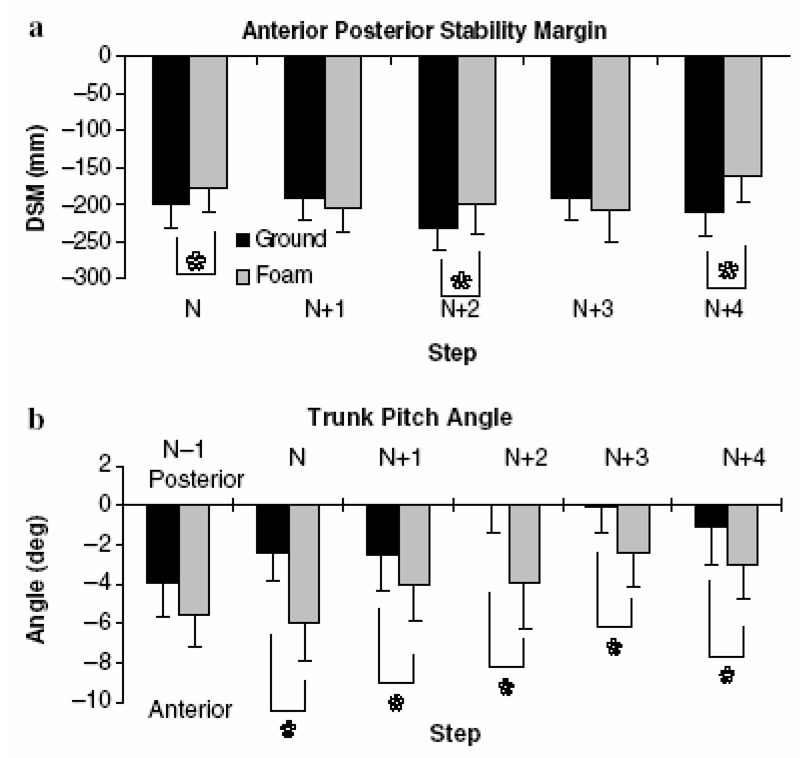


Fig. 7: **a** Stability margin in the anterior–posterior direction. **b** Estimated three-dimensional trunk pitch angles at each foot contact. Significant differences are denoted by *

a maximum observed at FC_{N+2}. No trial effects were observed in DSM_{AP} or DSM_{ML}.

3.3.7 - Three-dimensional trunk angles

During locomotion on the compliant surface, the trunk was significantly pitched more forward when compared to the ground condition. This can be seen in Fig. 7b. Significant trunk pitch was illustrated through an interaction effect ($F_{(5,33)}$=3.64, P<0.0098). No significant differences were seen in trunk roll or trunk yaw. There were no trial effects in either trunk pitch, roll, or yaw.

3.4.1 - Discussion

Strategies for stepping on a visible compliant surface

3.4.2 - Proactive control of the vertical center of mass

Analysis of the vertical COM revealed that there was an initial increase in vertical COM trajectory prior to stepping on the compliant surface. This initial increase represents a proactive attempt, to compensate at least partially, for the subsequent lowering of COM due to surface depression when stepping onto the compliant surface. Ferris et al. (1999) argue that vertical COM excursion is maintained when running on a compliant surface. A significant decrease in vertical COM following first contact on the compliant surface is not just due to the compression of the compliant surface: it is also an active strategy by the CNS to increase stability by lowering COM as seen during an expected step on a compliant surface (Marigold and Patla 2005) and in response to locomotion on a slippery surface (Marigold and Patla 2002). Lowering of the whole body COM decreases the moment arm between the COM and the ground reaction force, which means a greater amount of force would be needed in order for a fall to occur. Vertical COM continued to lower on the compliant surface over consecutive steps. This suggests a feedback mechanism is used in the lowering of the vertical COM as walking continued on the compliant surface, to minimize threats to stability.

Medio-lateral whole body COM control did not differ between the compliant surface condition and the ground condition. This suggests better active regulation of COM in the frontal plane during walking on a compliant surface. The increases in step width variability is evidence that constantly changing step width is used effectively to control center of pressure to maintain medio-lateral body COM within normal limits.

3.4.3 - Tripping on the compliant surface is avoided through increases in toe elevation during the swing phase

The toe trajectory profiles show greater initial maximum and minimum peaks when stepping onto the compliant surface. This represents a proactive response to ensure a larger safety margin between the compliant surface and the toe to minimize chances of an accidental trip. Similar increases in toe clearance were observed in Marigold and Patla (2005) when stepping off of a compliant surface to decrease the risk of tripping. When walking on the compliant surface, the height of the surface is constantly changing especially during toe off and foot contact. This increase in toe clearance would ensure enough clearance to avoid any toe contact with the changing compliant surface height during the swing phase. No trial effects were evident in toe elevation values suggesting that learning was not a factor.

3.4.4 - Stability is controlled through coordination of all body segments

Stepping onto an unstable surface can threaten stability during locomotion. When a surface change is clearly distinguished by color for example, as was the case in the present study, the visual system can pick up cues in the environment to be used for proactive changes in walking patterns to ensure stability. For example, step width and length increased in the initial step on the compliant surface, which were achieved by appropriate prior changes in muscle activation profiles. By increasing step width and length, a larger BOS is created in which the COM can be controlled. Step length changes occurred through increases in stance limb muscular activation prior to stepping onto the compliant surface. The increase in left gastrocnemius activity during prior left limb push off on the firm ground would increase the step length (Patla et al. 1989) which was larger than other steps on the compliant surface. This confirms the hypothesis that the CNS will collect information used to increase BOS to maximize stability. The increases seen in the step length and width variance suggest an increased need for regulating foot placement to continually deal with perturbations to stability.

Anterior–posterior stability margin increased on the compliant surface for the first step onto the compliant surface, even though the trunk was pitched forward during this step. Marigold and Patla (2005) also observed similar increases in trunk pitch when stepping onto an unknown compliant surface. Therefore, this increase in stability margin occurred due to increases in step length and/or other body segments pulling the full body COM backward. During the second step, anterior–posterior stability margin decreases, which may be due to the increase in trunk pitch seen in the second step on the compliant surface. Since these changes were made prior to stepping on the compliant surface, they represent proactive adjustments to locomotion to maintain stability. No significant differences were observed in COM, trunk roll, trunk yaw, and margin of stability in the medio-lateral direction. This result illustrates that proactive adjustments are more tightly regulated in the frontal plane when compared to sagittal and transverse planes. No trial effects were observed in stability margin or trunk orientation, suggesting that a cautious response to stepping on the altered surface, which can be modified following experience with the task, was not present.

Strategies for taking several steps on a visible compliant surface

3.4.5 - Vertical center of mass is controlled through reactive mechanisms when walking on a compliant surface

Vertical COM tends to decrease as walking continues on the compliant surface. This decrease does not concur with previous studies (Ferris et al. 1998), but may occur due to an adaptive decrease in vertical COM to increase stability. This adaptation occurs to accommodate the compliant surface and maintain stability throughout contact. A decrease in vertical COM was observed in the compliant surface condition over successive steps. This suggests a feed forward mechanism used to increase stability by lowering whole body COM as walking continues on the surface. There was also a phase lag seen in the compliant surface vertical COM trajectory when compared to the stable surface vertical COM trajectory. This phase lag would suggest that vertical COM is reactively controlled while walking on the foam. The CNS needs to gather information from foot contact on the compliant surface to reactively control vertical COM.

Again, while walking on the compliant surface, no changes are seen in medio-lateral COM peak amplitudes. This is similar to what is seen when participants initially step onto the compliant surface and the large increases in step variability may be in order to control center of pressure to maintain medio-lateral COM within normal limits.

3.4.6 - Base of support continues to increase while walking on a compliant surface

When walking on the visible compliant surface in this study, increases were seen in step width, length, and time. The increases seen in step length corresponded to findings by McMahon and Greene (1979). Step width and length increased to ensure an enlarged BOS. This would enhance stability. The increases in the step length agree with the increases of muscle activity in the gastrocnemius and soleus muscles during prior push-off. These changes are most likely proactive in order to better control the COM within the BOS. Step width and length variability were also increased while walking on the compliant surface. The greatest amount of step width variability was seen in FC_{N+2}, the first step where both feet are on the compliant surface for the first time. At this time, the increase in variability may be due to a problem in implementing the proactive strategy to increase BOS or it may be due to dealing with the variability in stability demands when walking on the compliant surface. Variability was increased in step length, which could be related to errors in estimations during proactive foot placement. Step time increased as participants walked on the compliant surface but analysis from step velocity showed that there was no overall decrease on the compliant surface. This shows that the increase in step time was due only to increases in the length of the step.

Stability margin in the anterior–posterior direction was greater in FC_{N+2} and FC_{N+4} on the compliant surface but less in FC_{N+3}. Since the step length is consistently larger in the compliant surface condition, changes in anterior–posterior stability margin in FC_{N+3} cannot be due to foot placement. Increased forward trunk pitch can account for the decrease in margin of stability for FC_{N+3}. These changes may occur due to overcompensation in correcting total body COM in the predicted step. The changes in stability margin over sequential steps show the evolution of the recovery strategy to perturbations to balance.

3.4.7 - Limitations

A major limitation in this study is the ability to determine what observations in the study are due to the sensory perturbation or the mechanical perturbation caused by the surface. Foam surfaces have been used previously to perturb the kinesthetic sensation in the foot. Along with this sensory perturbation, the viscoelastic nature of the foam surface causes a mechanical perturbation to gait through the deformation of the walking surface. In the present study, it is not possible to determine which observations occur due to the sensory perturbation, mechanical perturbation, or a combination of the two perturbations. Some studies have examined locomotion in participants with vibrating foot soles. This would also cause a sensory perturbation to the kinesthetic system. The problem with this is that vibrations may interfere with other components of the kinesthetic sensory system when compared to walking on a foam surface. This issue should be examined more extensively to determine how to separate observations due to sensory and mechanical perturbations.

When the anterior border of the BOS was determined in this study, markers placed on the 5th metatarsal were used. The use of these markers would actually underestimate the location of the anterior border of the BOS because the marker is placed more posteriorly on the foot when compared to where the actual location of the BOS (1st metatarsal). In this study, a marker could not be placed on the 1st metatarsal because it would be covered due to the foot's compression in the compliant surface. To correct for this, a virtual marker could be calibrated to the front of the 1st metatarsal so that accurate measurements can be made for the anterior border of the BOS.

3.4.8 - Future Directions

From the results in this study, it can be seen that locomotion on a compliant surface challenges the CNS and causes changes in normal walking pattern. It is well known that older adults do have problems when walking on compliant surfaces such as sand and grass and confidence decreases in these situations. Future studies should examine how older adults adapt to walking on a surface such as this and determine how their walking pattern changes during adaptation. In making comparisons to a younger population, it may be possible to determine if older adults are unable to adapt to specific aspects of the compliant surface and biomechanical reasons for decreases in confidence can be determined. From these results, rehabilitation programs could be implemented into retirement homes in order to increase the confidence in older adults with locomotion.

3.5.1 - Conclusion

In conclusion, when walking on a compliant surface, the CNS coordinates the whole body in order to maximize stability. Vertical COM is decreased proactively when stepping onto and as walking continues on a compliant surface. Medio-lateral COM does not change when walking on the compliant surface and this may occur due to a tight control by the CNS in the frontal plane. This control is achieved through changes in step width, which are the primary means of controlling COM. This agrees with the hypothesis stated previously. Increases in step length and time are actively controlled to increase stability. All of these changes occur to maximize stability when it is threatened due to surface changes.

4.1.1 – Stepping over an Obstacle on a Compliant Travel Surface Reveals Adaptive and Maladaptive Changes in Locomotion Patterns

MacLellan, M.J. & Patla, A.E. (2006). Stepping over an obstacle on a compliant travel surface reveals adaptive and maladaptive changes in locomotion patterns. *Exp Brain Res*. 173(3): 531-8.

Human locomotion depends heavily on making necessary proactive changes when obstacles arise in the travel path. Depending on the size of the object, one may decide to steer around the obstacle or step over it. Obstacle clearance is accomplished during single support of one limb and during swing phase in the other; the decreased base of support while the limb is elevated can pose a risk to balance. This task would clearly be more challenging when the surface one is walking on is irregular or compliant. This would put added demands on the central nervous system (CNS) to maintain stability. This study examines how obstacles are approached when walking on a compliant surface and how toe clearance is achieved. Locomotion on compliant surfaces can cause many neuromuscular adaptations. Surface changes alone can cause alterations in lower limb kinematics (Hardin et al. 2004), leg muscle activation (Ferris et al. 1999; Moritz and Farley 2004), and toe clearance in swing phase (Marigold and Patla 2005).

Step length modifications, toe elevation, and whole body center of mass (COM) are among the many factors the CNS must control to successfully avoid an obstacle during locomotion. This must be done to ensure appropriate foot placement during the approach phase and in the step over the obstacle. Lee et al. (1982) showed that foot placement variability decreases in long jumpers as they approach takeoff. Similar reductions in foot placement variability were observed by Patla and Greig (2005) when people were asked to step over an obstacle: appropriate variability in foot placement before the obstacle was a determining factor in successful obstacle clearance. Walking on a compliant surface increases step length variability in order to control anterior–posterior whole body COM (MacLellan and Patla 2006). Stability is dependent on controlling whole body center of mass (COM) within a constantly moving base of support: foot placement is the primary means of controlling stability (Patla 2003). To account for body movement, Hof et al. (2005) proposed a method in which the velocity of the whole body COM is used to adjust the COM position; instability occurs when the calculated value of dynamic stability margin (DSM) becomes less than zero. Since a systematic reduction in foot placement variability is a determining factor in successful obstacle clearance, the compliant surface poses competing demands of successful obstacle clearance and maintenance of stability.

A major factor in obstacle avoidance is toe trajectory over the obstacle: toe elevation must be sufficient to safely clear the obstacle. It has been documented that this clearance is about 0.01 m (Patla and Rietdyk 1993) and is achieved through coordination of ankle, knee, and hip joint kinematics and kinetics (Patla and Prentice 1995). During the elevation phase of obstacle clearance, research shows that knee power is increased to elevate the toe (Patla and Prentice 1995;

Niang and McFadyen 2004). During limb lowering after obstacle clearance, Patla and Prentice (1995) showed that work is absorbed at the hip joint and generated at the ankle to ensure a gentle landing after obstacle clearance. Walking on a compliant surface alone causes an increase in knee power to avoid tripping during toe off (Marigold and Patla 2005). How the two competing goals of maintaining stability on the compliant surface and obstacle clearance are satisfied by the CNS is not known. Compliant surfaces cause the lowering of the stance limb due to the depression of the surface. Does the CNS take into account this depression of the stance limb to regulate elevation such that toe clearance is maintained? Is the strategy to achieve limb elevation modified when walking on a compliant surface?

The main purpose of this study is to determine how the CNS organizes movement during approach to an obstacle and how safe toe clearance is achieved during locomotion on a compliant surface. Answers to these questions will shed light on how the CNS copes with two concurrent threats to stability.

4.2.1 - Methods

4.2.2 - Participants

Six participants (three female and three male; age 21±1.5 years; mass 70.4±15.4 kg) volunteered for this study. Participants had no muscular, neurological, or joint disorders which would affect their performance in this study. The study was approved by the Office of Research Ethics at the University of Waterloo and written informed consent was received from all participants.

4.2.3 - Compliant surface

A medium density foam (5 m long, 0.91 m wide, 0.12 m deep) was used as the compliant walking surface in this study. An elevated wooden platform (1.1 m long, 1.2 m wide, 0.12 m deep) was used as a starting position during the compliant surface condition in order to avoid stepping up onto the surface. A diagram of the setup can be seen in Fig. 1a.

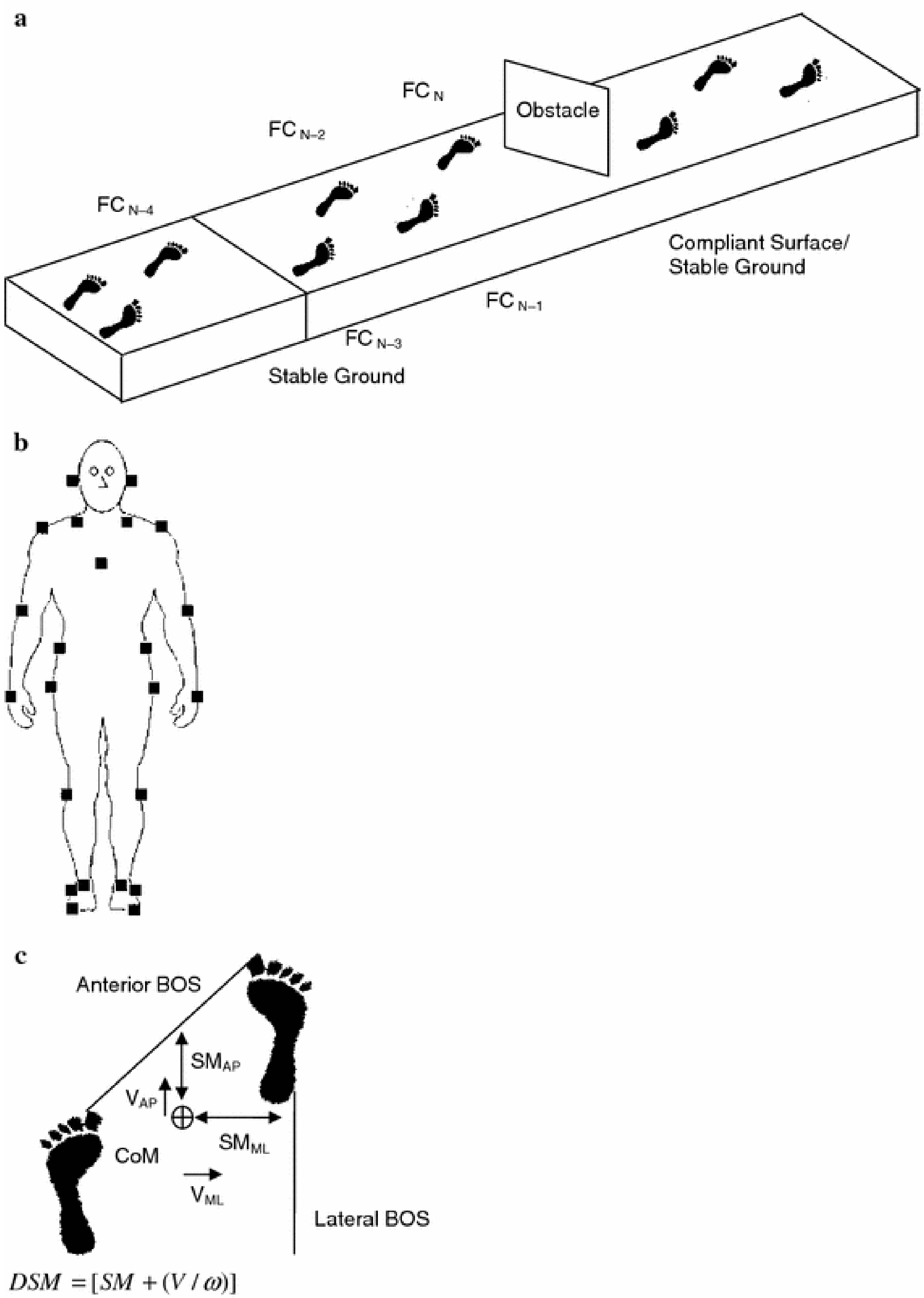


Fig. 1: Experimental setup. **a** Identification of steps prior to obstacle clearance. **b** iRED placements for Optotrak measurements. **c** Calculation used to determine dynamic stability margin

4.2.3 - Protocol

Two blocks of walking trials were collected: 20 baseline ground walking trials and 20 compliant surface walking trials. These 20 trials were split into two blocks of ten trials (ten obstacle and ten no obstacle). The order of obstacle and no obstacle blocks were randomized. Baseline ground walking was completed first by all participants. During the baseline condition, thin lengths of green mat were placed on the ground to keep the walking area constant between the two conditions. Participants were instructed to start walking at a normal pace initiating gait with the left foot. Compliant surface trials were collected next. The obstacle was a thin piece of wood (0.3×0.7×0.005 m) that was placed at five distances on the walkway (approximately 3.095, 3.215, 3.335, 3.455, and 3.575 m from start position). Each position was presented two times in a random order.

Whole body kinematics were measured using three Optotrak cameras (Northern Digital, Inc., Waterloo, Canada) at a frequency of 80 Hz. Twenty-three iREDs were placed on the following anatomical land marks: fifth metatarsal, top of ankle, lateral malleolus, fibular head, greater trochanter, iliac crest, clavicle, acromion process, olecranon process, radial styloid process, and xiphoid process bilaterally (seen in Fig. 1b). This allowed us to determine lower limb kinematics, kinetics during swing, and three-dimensional trunk positions.

4.2.4 - Data analysis

All kinematic data from iRED markers were low pass filtered at 7 Hz (dual pass, second order, Butterworth filter). Ankle markers were used to determine step length. Steps were termed in relation to obstacle placement. The initial step that participants took with their left foot was termed FC_{N-4}, and the following step with the right foot was termed FC_{N-3} with FC_{N-2}, FC_{N-1}, and FC_{N} following (see Fig. 1a). From these steps, distance from the step to the obstacle (foot placement) was determined. Toe trajectory was determined for each participant using the iRED placed on the fifth metatarsal. Ankle, knee, and hip angle trajectories were determined using methods proposed in Winter (2005). From these trajectories, toe elevation, ankle angle, knee angle, and hip angle at the obstacle were determined. Using three trunk markers (right clavicle, left clavicle, and xiphoid) true three-dimensional trunk angles were determined at the obstacle using the Cardan method of determining angles. Stability margin was calculated as in Fig. 1c. Stability margin in the anterior–posterior direction (SM_{AP}) was measured as the distance between the line joining the toes to the COM at heel contact. Stability margin in the medio-lateral direction (SM_{ML}) was measured from the lateral malleolus of the leading foot to the COM at heel contact. This measurement was calculated for each heel contact. To determine the effect of COM velocity on stability margin, termed DSM, the position of the COM was added to the estimated instantaneous velocity of the COM divided by instantaneous angular velocity of the COM as in Hof et al. (2005). Swing limb moments, powers, and work at the ankle, knee, and hip were determined using methods proposed in Winter (2005). Swing limb joint moments and powers were determined at the obstacle. For analysis of work, work for each joint was determined for ground/compliant and obstacle/no obstacle trials in two phases. Phase 1 occurred from toe off to half of toe trajectory and phase 2 from half of toe trajectory to heel contact.

4.2.5 - Statistical analysis

A condition (ground, compliant) × step (FC_{N-4}, FC_{N-3}, FC_{N-2}, FC_{N-1}, FC_{N}) analysis of variance (ANOVA) was administered on step length, step length variance, foot placement, and foot placement variability. A surface (ground, compliant) × obstacle (no obstacle, obstacle) ANOVA was administered to determine any differences in DSM and joint power. A student's t test was administered an all other measures to determine significant differences between the ground and compliant surface conditions. For all analysis, an alpha level of 0.05 was used to determine significance. Post hoc analysis was conducted when significant differences existed using a Least Squares Difference test with an alpha level of 0.05.

4.3.1 - Results

4.3.2 - Step measurements

Step length and foot placement did not differ between the ground and compliant surface conditions. ANOVA determined that there were no significant differences between the ground and compliant surface condition step lengths ($F_{(3,18)}$=0.39, P=0.7609). There were also no differences determined in foot placement between the conditions ($F_{(4,20)}$=0.48, P=0.7487). There were significant differences when examining the variability of step length. A condition effect was seen in step length variance in which compliant surface step variability was greater than ground step variability ($F_{(1,5)}$=15.03, P<0.0117). There was also a step effect in which variability increased as steps progressed ($F_{(3,15)}$=9.17, P<0.0011). When examining foot placement variability, a step effect was observed where variability decreased as steps approached the obstacle ($F_{(4,20)}$=13.49, P<0.0001; seen in Fig. 2).

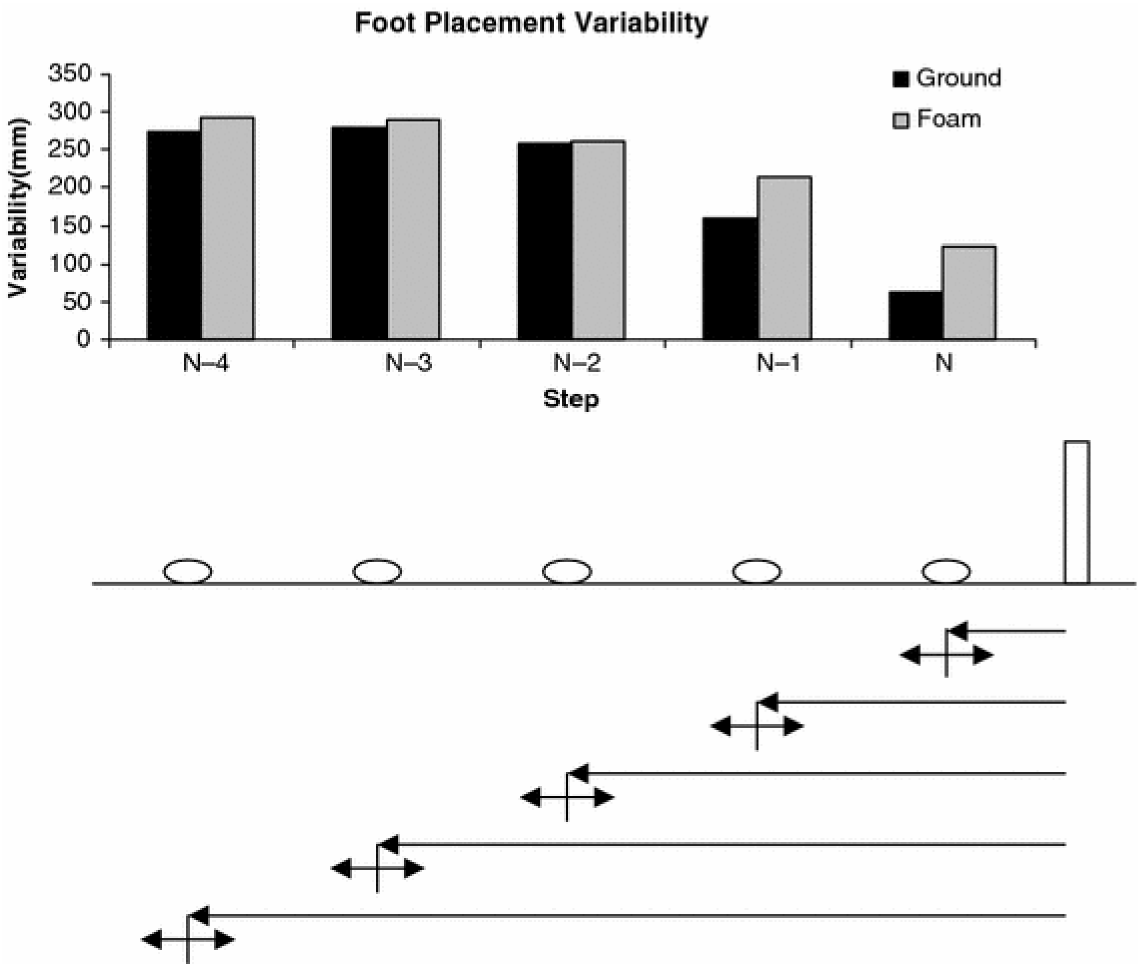


Fig. 2: Foot placement variability in steps prior to obstacle clearance. *Bottom* illustrates how variability measurements were obtained

4.3.3 - Lower limb kinematics

Toe elevation (distance from toe off to trajectory at obstacle clearance) at the obstacle was not different between the two conditions. Toe trajectories can be seen in Fig. 3a. Mean toe elevations were 0.451 (±0.061 m) and 0.459 m (±0.048 m) for ground and compliant surface conditions, respectively. A student's t test determined that toe elevation was not different between the ground and compliant surface conditions ($t_{(112)}=0.79$, $P=0.2163$). Toe clearance (distance from top of obstacle to trajectory at obstacle clearance) at the obstacle was different between the ground and compliant surface conditions. Mean toe clearance was 0.161 (±0.018 m) and 0.150 m (±0.013 m) for ground and compliant surface conditions, respectively (seen in Fig. 3b). A student's t test determined that toe clearance was significantly higher in the ground condition ($P<0.0151$) and that toe elevation variance was significantly smaller in the compliant surface condition ($P<0.0079$).

Increases in knee and hip flexion are seen during toe elevation in the swing limb. At obstacle clearance, no differences were observed in ankle angles ($t_{(111)}=-0.34$, $P=0.3674$) but differences were observed in knee ($t_{(111)}=1.87$, $P<0.0321$) and hip ($t_{(111)}=-4.23$, $P<0.0001$) angles. Knee flexion at obstacle clearance was 86.21±6.7 and 89.44±8.8 degrees for ground and compliant surface conditions, respectively. Hip flexion was 63.78±3.8 degrees in the ground condition and 70.41±3.1 degrees for the compliant surface condition. No

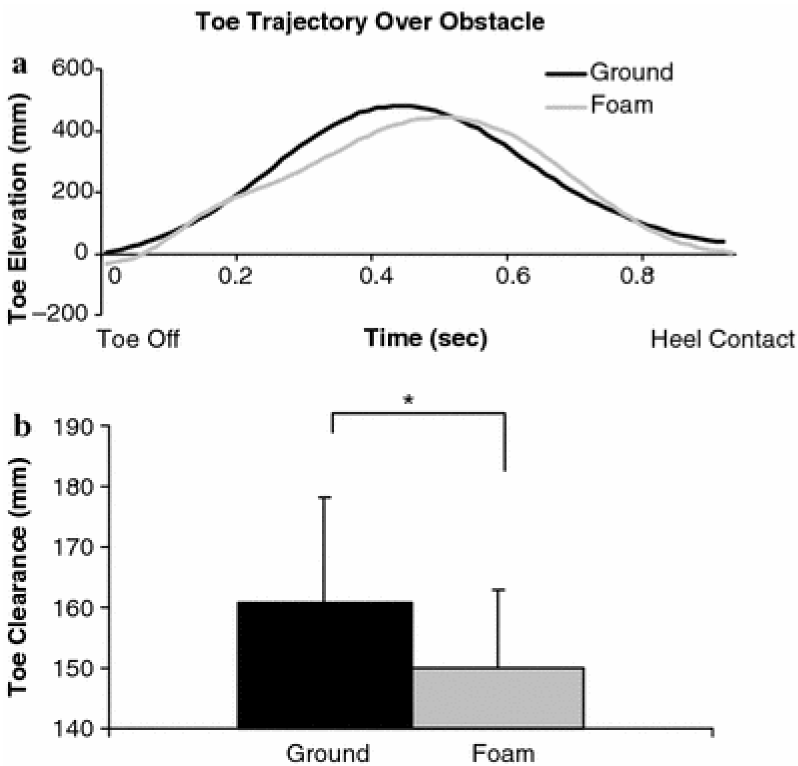


Fig. 3: **a** Representative trials of foot trajectory during obstacle clearance. **b** Toe clearance (measured from top of obstacle to trajectory at obstacle clearance) for ground and compliant conditions. *Asterisk* indicates significance of <0.05

differences were seen in trunk pitch ($T_{(112)}$=0.73, P=0.234) or roll ($T_{(114)}$=−0.51, P=0.307) in the ground or compliant surface conditions. Trunk pitch and roll at obstacle clearance was 5.57±3.5 degrees forward and 0.856±2.6 degrees left in the ground condition and 5.94±3.8 degrees forward and 0.243±2.9 degrees left in the compliant surface condition.

4.3.4 - Dynamic stability margin

DSM_{AP} increased as participants approached the obstacle. DSM_{AP} is illustrated in Fig. 4. A significant obstacle by condition by step interaction was seen in DSM_{AP} (F(4,18)=3.67, P < 0.0224) but no significant differences were seen in DSM_{ML} (F(4,18)=1.84, P < 0.1620) DSM_{AP} tends to increase as the participant approached the obstacle, but this was similar between the ground and compliant surface conditions.

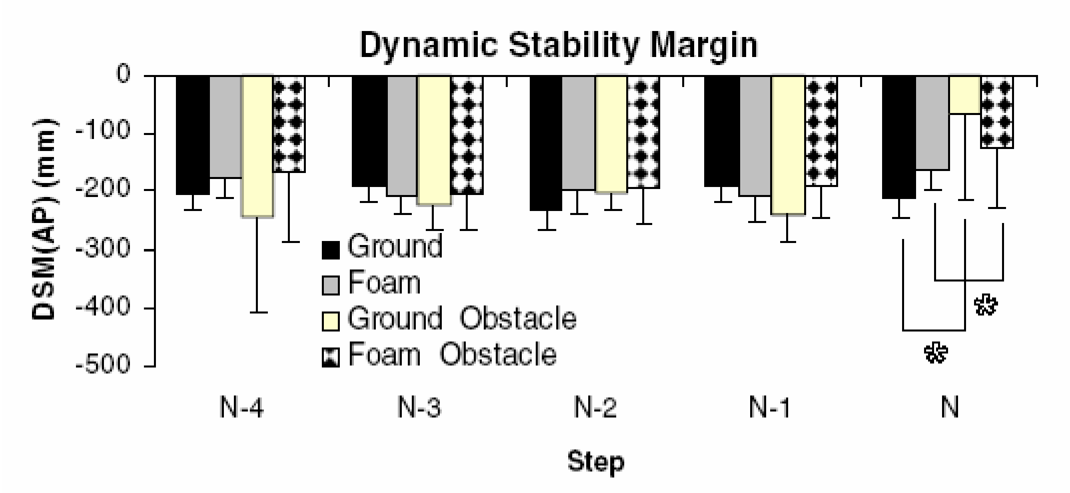


Fig. 4: Plot of dynamic stability margin during the approach to the obstacle. *Asterisk* indicates significance of <0.05

4.3.5 - Lower limb kinetics

An increased flexor moment in the ankle was seen at obstacle clearance in the compliant surface condition. Ankle flexor moment increased from 0.805±0.103 Nm on the ground to 0.933±0.110 Nm on the compliant surface ($t_{(111)}=2.69$, $P<0.0041$). No significant differences were observed in knee moment ($t_{(111)}=-0.34$, $P=0.367$) or hip moment ($t_{(111)}=-0.37$, $P=0.356$). Ankle ($t_{(111)}=2.56$, $P<0.0059$) and knee ($t_{(111)}=3.24$, $P<0.00079$) power was significantly more negative at obstacle clearance. Power profiles for ankle, knee, and hip are illustrated in Fig. 5. Ankle power decreased from −0.659±0.549 W on the ground to −1.095±0.667 W on the compliant surface. Knee power decreased from −12.539±3.456 to −16.087±4.326 W from the ground to compliant surface conditions. No significant differences were observed in the hip power at obstacle clearance ($t_{(111)}=-1.13$, $P=0.13$). Work done at each joint from toe off to obstacle clearance and obstacle clearance to heel contact can be seen in Table 1. It was determined that there were no significant differences observed in ankle work during phase 1 ($F_{(1,5)}=2.20$, $P=0.198$) or phase 2 ($F_{(1,5)}=0.54$, $P=0.494$). A condition effect was observed in knee work during phase 1 ($F_{(1,5)}=66.04$, $P<0.0005$) in which work was more negative in the obstacle condition. No differences were seen in the knee joint during phase 2 ($F_{(1,5)}=0.26$, $P=0.632$). When observing the hip joint, no differences were seen in phase 1 ($F_{(1,5)}=0.31$, $P=0.601$) but a obstacle by surface interaction effect was observed in phase 2 ($F_{(1,5)}=11.78$, $P<0.0186$). This interaction effect illustrated that work was greater in

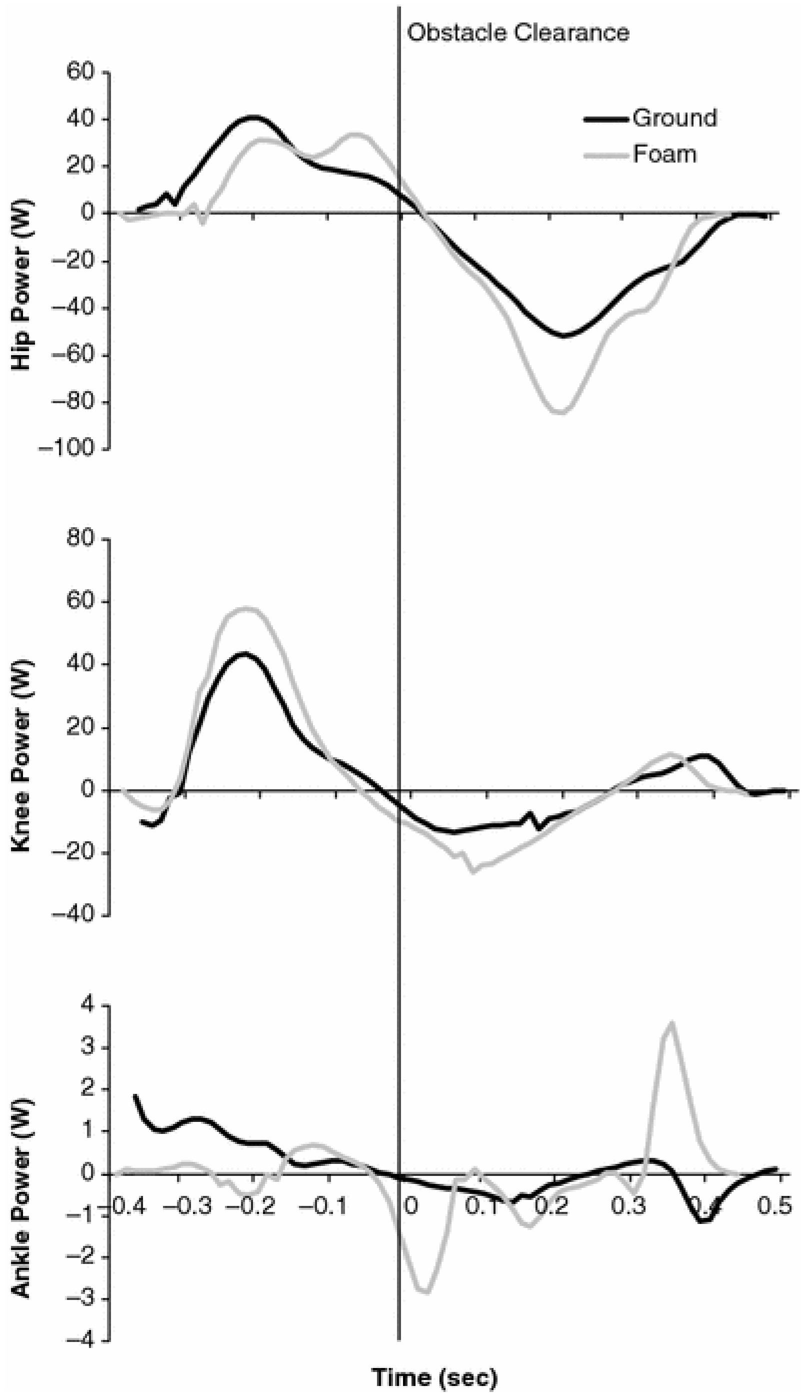


Fig. 5: Representative trials of estimated joint power for hip (*top*), knee (*middle*), and ankle (*bottom*). *Positive values* indicate power generation and *negative* absorption

Table 1: Joint work (J) during obstacle clearance

Table 1 Joint work (J) during obstacle clearance

Joint	Phase	Ground/no obstacle	Ground/obstacle	Compliant/no obstacle	Compliant/obstacle
Hip	Elevation	2.44639	6.1657	2.2501	5.4076
	Lowering	−4.0185	−19.6146	−6.8639	−19.3233
Knee	Elevation	1.5732	9.0599	1.3699	10.2211
	Lowering	−4.7952	−3.2218	−5.6289	−4.5682
Ankle	Elevation	0.0444	0.0517	1.0211	0.1208
	Lowering	0.1269	3.7126	−1.9785	−1.3527

the compliant surface condition when compared to the ground condition when no object was present and that power was significantly greater in the object condition when compared to the no object condition.

4.4.1 - Discussion

In this study, two concurrent threats are presented to the CNS: a stability threat due to the compliant surface and clearance of an obstacle in the travel path. How the CNS deals with these competing stability threats was assessed by analyzing whole body kinematics and lower limb kinetics. Successful obstacle avoidance is dependent on appropriate foot placement adjustments during the approach phase (Patla and Greig 2005) and adequate toe elevation to clear the obstacle (Patla et al. 1991). While the CNS is able to appropriately adjust foot placement in the approach phase over a compliant travel surface, obstacle clearance shows maladaptive changes that can potentially threaten stability.

4.4.2 - Competing demands of maintaining dynamic stability and foot placement before the obstacle are achieved

In a study of elite long jumpers, Lee et al. (1982) observed that standard error of footfall position decreased as the jumper approached a takeoff board. Similar results were reported by Patla and Greig (2005) in foot placement variability when participants approached an obstacle in their travel path. This decrease in variance illustrates an increase in control by the CNS of the final foot placement before stepping over the obstacle. Results from this study suggest foot placement, not step length is regulated to prepare for obstacle clearance. These results contrast what was seen by Begg et al. (1998) where step length increased prior to obstacle clearance. The control of foot placement before the obstacle is clearly important to minimize the risk of tripping in the following step. Researchers (Chou and Draganich 1998; Patla and Greig 2005) argue that the appropriate placement of this step determines the trajectory of the foot while stepping over the obstacle. Similar decreases in foot placement variability were observed in this study, irrespective of whether the individual was walking on a compliant or normal surface. This illustrates that the CNS is able to make the specific foot placement adjustments needed to prepare for obstacle clearance while walking on a compliant surface.

Foot placement control is also needed to control stability. MacLellan and Patla (2006) concluded DSM_{AP} was significantly different when walking on a complaint surface. This study showed continuous over compensatory control of DSM_{AP} during locomotion on a compliant surface. It is possible that stability may be challenged when approaching an obstacle. Patla and Greig (2005) and the current study illustrates that foot placement is altered to decrease foot placement variability when approaching an obstacle. Since dynamic stability is dependent on foot placement, this change in variability may have an affect on dynamic stability. The results show that dynamic stability is maintained while approaching an obstacle on a compliant surface. Analysis of DSM_{AP} illustrates no differences in stability during the initial steps when approaching an obstacle. The second to last step prior to obstacle clearance, DSM_{AP} increases from the compliant/no obstacle to the compliant/obstacle conditions. In the final step prior to obstacle clearance, DSM_{AP} increases (primarily due to adjustments of COM position) in the obstacle condition compared to the no obstacle conditions on both ground and compliant surfaces. This indicates that the CNS initiates stability changes in steps prior to obstacle clearance irrespective of the travel surface to minimize the chances of a fall should an accidental trip occur. These results illustrate the competing demands of stability and foot placement are met on the compliant surface.

4.4.3 - Strategies for toe elevation and lowering are similar for compliant and normal terrains

Two major strategies have been documented to achieve successful obstacle clearance: knee and hip strategies. Patla and Prentice (1995) showed that the muscles about the knee joint are responsible for toe elevation. In this study, knee power to obstacle increased as a function of obstacle height. A different strategy was observed by Hill et al. (1999) in lower limb amputees in which hip work was modulated as obstacle height increased. In people with lower limb amputations, two challenges may exist: a reduction of kinesthetic input and an increase of joint instability. These challenges are also present during locomotion on a compliant surface. Nevertheless, results from this study show that on a compliant surface, a similar knee strategy is used to elevate the toe during obstacle clearance.

A hip strategy is used to lower the limb after obstacle clearance on the compliant surface. Patla and Prentice (1995) concluded that hip power absorption increased as obstacle height increased during the limb lowering phase. Absorption of hip power was similar in the ground and compliant conditions, which corresponds to using a hip strategy to lower the limb after obstacle clearance. In the compliant surface condition, the work done by the hip is similar even when toe clearance is lower over the obstacle. This shows that the same amount of power is absorbed over differing lowering distances. This greater absorption of work in the compliant surface condition functions to decrease vertical contact velocity. The compliant surface has viscoelastic properties which causes unpredictable weight bearing surface deformation during contact. An increased vertical velocity would be detrimental on the compliant surface because the compression of the surface could compromise stability.

4.4.4 - The CNS regulates toe elevation, not toe clearance when stepping over obstacles

Stepping over an obstacle can be a hazardous task due to the risk of tripping during clearance. Thus, sufficient elevation of the foot must ensue to safely clear the obstacle (Patla et al. 1991). Patla and Prentice (1995) have illustrated toe elevation is achieved by coordinating joint positions of the ankle, knee, and hip. On normal stable ground, toe elevation and toe clearance are related; researchers have assumed that the CNS is regulating certain toe clearance to ensure safe travel (Patla et al. 1991). Results of this study suggest that toe elevation, not toe clearance is controlled when stepping over an obstacle because the depression of the surface during weight bearing is not taken into account; toe clearance over the compliant surface is smaller. This decrease in toe clearance would increase the chances of contact with the obstacle resulting in a trip possibly leading to a fall. This is an important finding indicating CNS function during obstacle clearance. In previous studies, toe elevation and clearance have been coupled. In this study, the obstacle height is maintained, providing a consistent visual perception of the obstacle but the ground the obstacle is on compresses causing a change in toe elevation needed to successfully clear the obstacle. These results indicate the CNS either does not take into account the compression of the ground prior to obstacle clearance or due to the unpredictable nature of the compliant surface, it does not have adequate information to accurately judge the compression of the surface during obstacle clearance.

Whatever the reason, the inability of the CNS to estimate surface compression and limb lowering is reflected in similar work done during limb elevation. While the visual system provides the estimate for the work needed to elevate the toe during obstacle clearance, it is not able to take into account any compression of the surface prior to clearance and adjust the motor patterns accordingly. If the CNS was able to estimate compression of the ground prior to obstacle clearance, differences would be seen in joint work to compensate for ground compression.

4.4.5 - Limitations

Similar to the previous study, the compliant surface used creates some limitations due to the viscoelastic properties of the foam. The foam surface creates sensory and mechanical perturbations to the CNS therefore it is not possible to determine what changes occur due what aspects of the surface. This is one limitation occurring in this study.

Along with the surface characteristics, this study only used one height of obstacle for avoidance. It is possible that the observations in this study will not be seen in obstacles of differing heights. The obstacle used in this study (0.3 m in height) is quite high and it is possible the CNS may use different strategies to cross obstacles of lower heights. To examine this, a study can be done to examine toe clearance and elevation during obstacle avoidance with obstacles of differing heights and possibly widths.

A final limitation seen in this study is the marker that was used in determining toe elevations and clearance. The marker placed on the 5th metatarsal was used in this calculation. During clearance, the part of the foot most likely to make contact with an obstacle is the 1st or 2nd metatarsal. An issue created by walking on the compliant surface was that markers placed on the 1st or 2nd metatarsals would be covered by the surface itself and therefore the cameras would not be able to record their movements. To correct for this, a virtual marker could have been calibrated at the 2nd metatarsal and more accurate measurements could be made for toe clearance and elevation.

4.4.6 - Future Directions

To address the limitations presented, a new study has been proposed in which obstacles of differing heights will be used along with differing combinations of surfaces. By using obstacles of differing heights, it can be determined if toe elevation and clearance strategies are similar during avoidance. The surface combinations used in this study were ground and ground-foam. Along with these, foam and foam-ground surface combinations should be used. These combinations will confirm the conclusions of this study concerning the CNS made estimations about limb elevation prior to stepping on the foam surface. If foam only surface is used, toe elevation would be larger over the obstacle because estimations of limb elevation would be made while the participant is standing on the compressed foam. In the foam-ground condition, limb elevation would be larger than the ground and foam surfaces because the CNS would make estimations about limb elevation while standing on the compressed foam, and this would cause an overestimation of the elevation needed to safely clear the obstacle. These issues should be addressed in a follow-up study.

4.5.1 - Conclusions

In conclusion, not all alterations by the CNS are advantageous during obstacle clearance when traveling over a compliant surface. When approaching an obstacle, foot placement variability decreases to ensure appropriate foot placement prior to obstacle clearance. Hip work is absorbed to decrease vertical foot velocity upon foot contact on the compliant surface. These alterations by the CNS are advantageous to maintain stability. Toe clearance is decreased in the compliant surface condition which would increase the chances of contact with the obstacle. These changes are detrimental to stability during obstacle clearance.

5.1.1 - Final Conclusions

Evidence from these studies shows that when examining dynamic stability; kinematic, kinetic, and electromyographical aspects of human movement should be integrated to determine how the CNS maintains balance. The studies in this thesis have used many biomechanical variables to examine reactive, predictive, and anticipatory control of dynamic stability. It can be seen that different aspects of stability can be controlled using differing mechanisms.

Reactive control of stability was seen when recovering from a frontal plane perturbation and during vertical COM decreases on the compliant surface. During a frontal plane surface translation, the CNS reactively directs full body COM in a direction opposite to the direction of the perturbation. This is accomplished through an extension strategy in the stance limb. Following the perturbation, the CNS reactively overcompensates COM-BOS distance in order to regain a normal walking pattern. When walking on a compliant surface, vertical COM is decreased reactively in order to increase stability.

Examples of predictive control of stability were seen in changes of foot placement and electromyographical patterns when walking on the compliant surface and by estimating limb elevation during obstacle clearance on a compliant surface. When walking on a compliant surface, the CNS prepares for this perturbation by taking wider and longer steps therefore increasing the BOS.

This is accomplished through increases in lower limb extensor muscles during push-off in walking. Evidence from this research also suggests the CNS predicts how high a limb needs to be elevated to safely clear the obstacle by predicting the required height. This is seen through lower limb kinetics and toe elevation being similar on ground and compliant surfaces.

Lastly, initial increases in vertical COM, increases in toe clearance during walking on the compliant surface, and decreases in foot placement variability when approaching an obstacle are examples of anticipatory control of stability. When preparing for a step onto a compliant surface, the CNS anticipates the lowering that will occur due to the compression of the surface and therefore an increase in vertical COM peak prior to stepping onto a compliant surface. This increase in vertical COM is accompanied with an increase in toe clearance to decrease the risk of the toe making contact with the surface during swing phase. When approaching an obstacle on a compliant surface, the CNS anticipates the crossing and a decrease in foot placement variability is observed in order to accurately plant the foot prior to obstacle clearance.

From a thorough examination of stability in different complex environments, it can be seen that dynamic stability can not be determine through a single measure. There are many factors in dynamic stability that work together to achieve a common goal of maintaining balance. To determine how the CNS is able to control stability through reactive, predictive, and anticipatory mechanisms, many biomechanical factors must be integrated to determine how this common goal of maintaining balance is met.

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