ASSESSMENT OF BALANCE AND COORDINATION WHILE WALKING ON SURFACES THAT EMULATE OUTDOOR TERRAINS

by

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A Thesis submitted to the Faculty of Graduate Studies of

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Archana Arun

ABSTRACT

The purpose of this study is to evaluate the balance requirements of steady-state walking on irregular/compliant surfaces that would emulate out-doors terrains. The hypothesis of the study was that during walking, stepping onto an irregular and/or compliant surface will cause mechanical disturbances which will be corrected at the level of foot and ankle and will not result in significant change in the whole body centre of mass and trunk acceleration signals. Twenty subjects, aged 20-40 were recruited for the study. Forty walking trials were collected on 6 different surfaces (1 control and 5 experimental) placed one at a time underneath a 5m carpet. The following variables were analyzed - Spatiotemporal gait parameters, peak plantar pressure and pressure time integrals, maximum excursion of centre of mass, maximum excursion of centre of foot pressure, cross correlation of centre of mass and ankle angle and the cross correlation between trunk and leg acceleration signals. Results showed that there was a significant change in the spatiotemporal gait parameters. This suggested that the experimental surfaces did create a challenging environment in terms of modification of gait patterns. Alterations in the plantar pressure parameters support the hypothesis that the simulated outdoor surfaces did produce significant changes in the foot to surface interaction. The results of the correlation analysis between ankle angle and the centre of mass indicated that subjects were less stable on the experimental surfaces as compared to the control surface. There were no significant changes observed in the trunk - shank acceleration signals and in the centre of mass for different surfaces which indicated that the disturbances were absorbed at the level of the foot and were not transmitted to the upper body or contributed to change in body coordination. The simulated outdoor surfaces thus made, proved to be effective in creating small disturbances and challenging balance control during the steady state of walking.

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1. INTRODUCTION

Good mobility of joints is a prerequisite for all instrumental activities of daily living such as walking outdoors, taking transportation, crossing the street etc. (Ferrucci et al 1991). Loss of mobility can occur due to events such as stroke, multiple sclerosis or hip/knee osteoarthritis and subsequent joint replacement surgeries. It can also happen by multiple predisposing factors such as the gradual decline of neurological or physiological fitness (Rubenstein et al 1994). Loss of mobility can result in decreased physical function, disability, and reduced quality of life. Fear of falling and loss of confidence can also lead to inactivity and further functional decline, depression, feelings of helplessness and social isolation. Weakness and walking problems are the most common causes of balance impairments and falls (For reviews see Wagenaar 2002). A fall is defined as an incidence in which a person involuntarily comes to rest on the ground or another lower level. It can be due to a major intrinsic event such as stroke or an overwhelming external hazard (Tinetti 1986). Fall injuries are a common problem among the older population and reduce their mobility skills by limiting function. Studies have shown that the elderly show an increase in the postural sway in standing position when compared with a younger population (Cunningham 1982). The deteriorating body systems such as vestibular, vision and somato-sensory further add to their problems and create a vicious cycle of further decreasing their mobility. Poor balance skills due to the above reasons make this group very vulnerable to falls. Even a small disturbance can result in a fall and sometimes in serious injury such as a hip fracture (Grabiner et al 1993). Researchers have advocated that regular out door walking is beneficial for maintaining good cardiovascular fitness and prevents the general health decline in older adults (Simonsick et al 2005).

2. LITERATURE REVIEW

2.1 BALANCE CONTROL

Balance has been defined as maintaining the position and velocity of the centre of mass (COM) in relation to the base of support (BOS) in all tasks such as standing, walking or running (Pai and Patton 1997). Pai et al (1997) studied an inverted pendulum model and found that to maintain balance, a moving body should have enough momentum to re-position the COM over the area circumscribed by the feet that are in contact with the ground i.e. the BOS or vice-versa. They demonstrated that to prevent falls during the movement termination, the COM velocity should fall within and around the location of the BOS. The walking task is an integration of balance as well as motor coordination. The stability of the body during such a dynamic task is a result of the interaction of several forces such as, gravitational, motion dependent, ground reaction forces and muscle activities (Johansson and Magnussson 1991). Balance during walking deserves special attention because of the complexities associated with this task. Dynamics of outdoor walking are different from standing because of the relationship of this task to the constantly changing and unpredictable environment. Apart from maintaining a proper relationship of COM position and velocity with the BOS, the central nervous system (CNS) also has to accommodate changes due to the alteration of surfaces or light conditions.

2.2 THE SENSORY ASPECT OF BALANCE CONTROL

Balance control is the outcome of a relative contribution of all sensory systems namely, vision, vestibular, proprioceptors in muscles and joints and cutaneous sensors in skin (Kooij et al 2001; Creath et al 2002; Peterka 2002). Each sensory component has a unique contribution. The vestibular receptors detect head position relative to gravity. The visual system can provide motion information and vertical location but has a relative frame of reference and thus sensory conflicts or illusions can arise (Kandel et al 2000). The proprioceptive component of the somatosensory system (i.e. muscle spindles and ligaments) provides information regarding relative orientation and motion of the body with the environment (Sargant 2000). The cutaneous afferents of the feet supply information about external reference points such as location, type of support surface and distribution of foot pressure. The cutaneous sensors are important in realizing the characteristics of surfaces we stand or walk on (Kandel et al 2000). Under normal conditions all these inputs are integrated for the correct response which is crucial in maintaining balance within the variety of environments encountered in daily life (Teasdale et al 1991).

However, if one of the inputs is diminished or absent, remaining components partially compensate for the loss (Allum et al 1998). The problem arises when there is an absence of two or more components. An inability to organize sensory information appropriately can result in instability in environments when visual cues are diminished or misleading (darkness, lack of contrast/depth cues), or the surface is unstable or compliant (sandy or snowy grounds or uneven surfaces such as gravel on side walks). An abnormality in these sensory inputs will lead to abnormal postural responses and hence balance disturbances; for example, low light or visual conflict situations exert more challenge for the CNS to maintain balance, and thus cause an increase in body sway (Brook-Wavell 2002). Similarly an irregular or compliant surface will also pose a postural threat by altering the somatosensory inputs from the foot (Redfern et al 1997).

The two important clinical tests developed to recognize multi-sensory organization and to quantify body sway and instability in the stance position are the Sensory Organization Test (SOT) and the Clinical Test of Sensory Interaction and Balance (CTSIB). Both these tests challenge the balance act by distorting the sensory information one by one and thus help identify the contribution of each sensor. The SOT was developed by Nashner in 1971 which utilizes a specialized tool called '*The Equi Test*'. This apparatus is comprised of a moving platform and a three sided surrounding visual cover (Nashner 1971, Peterka and Black 1990). The platform can rotate about an axis, co-linear with the ankle joint. Force plate recordings are used to measure the center of foot pressure. During the SOT, the sensory information delivered to the patient's eyes, feet and joints is effectively distorted through six different calibrated 'sway referencing' conditions of the support surface and/or visual surrounding, and thus creates sensory conflict situations. Subjects being tested can either cope with the loss of one or two sensory inputs, or can show some inaccurate postural responses (Di Fabio 1996). The six sensory conditions of the SOT protocol is comprises of:

Sensory Condition 1: The subject stands on a firm stable surface with eyes open.

Sensory Condition 2: The subject stands on a firm stable surface with eyes closed. Information from the visual source is thus eliminated.

Sensory Condition 3: The subject stands on a firm stable surface, and the visual surrounding sways in proportion to the body sway (visual-sway referencing). Here there is no relative motion between the head and visual surrounding but the body sways back and forth. The visual information is misleading or in conflict with this reality. In this case, if the subject uses only visual cues, then he/she will fall. The vestibular and somatosensory cues could provide accurate and functionally appropriate information to overcome the visual conflict.

Sensory Condition 4: The support surface on which the subject is standing is swayed. As subjects sway, the platform also rotates forwards or backwards to null the displacement about the ankle joint (*sway referenced support*). Here the somatosensory inputs provide distorted information about body sway. Vision and vestibular cues could provide accurate information in this condition and neutralize the platform action.

Sensory Condition 5: The subject stands on a *sway referenced support* surface with eyes closed. Here the somatosensory inputs are distorted and visual inputs eliminated. Vestibular inputs could provide accurate information.

Sensory Condition 6: The subject stands on sway referenced support surface. The visual surround is also sway referenced. Here the visual information is conflicting and somatosensory information distorted. The vestibular information is still accurate.

Performance during the SOT is quantified by the maximum peak-to-peak displacement of the centre of foot pressure in a 20 second time period. In young healthy individuals, for sensory conditions 1 and 2 there are only a few degrees of body sway. The body-sway progressively increases through conditions 3 to 6, as established from performance measures. Sensory conditions 5 and 6 are the most difficult conditions. The balance disturbance thus progresses from altering one sensory input, to distorting information from two sensors. Effective use of the sensory inputs is determined from the comprehensive scores on the six conditions. When a normal person is exposed to these six conditions the body-sway gradually increases, but does not result in loss of balance or fall, whereas, the balance of a person with a compromised sensory system may be challenged. The sensitivity and specificity of the SOT has been tested in vestibular disorders with vestibular function tests set as the criterion validity (Di Fabio 1996). The elderly population also has significantly increased sway as compared to the younger age groups in all the conditions of the SOT (Whipple et al 1993). Riley et al (2003) further expanded the analysis of the SOT and found that as the conditions of the test progressed, the quantity and variability of postural sway increased.

Based on the concept of the SOT Shumway-Cook and Horak in1986 devised a low-cost tool to test the sensory component of balance: the Clinical Test of Sensory Interaction and Balance (CTSIB). They used a medium density sponge with dimensions of 50×50×8 cm. The sponge replicated the SOT in terms of a *sway referenced support* surface with an added advantage that it was not limited to pitch plane but disturbance could be multi-directional. Also, the sponge being a compliant surface distorts somatosensory inputs for receptors of the foot. This technique uses three visual and support surface conditions, each tested for 30 seconds. The sway is quantified using : (i) a numeric ranking system (1 = minimal sway 2= mild sway 3= moderate sway and 4= fall), (ii) use of a stop watch to record the amount of time the patient maintains erect standing in each condition, and (iii) use of grids or a plum line to record body displacement. Compliant surfaces such as foam pads have become popular in producing balance disturbances in stance (Redfern 1997). Such compliant surfaces have an advantage in that they can cause perturbations in multiple directions in an unpredictable manner. The sponge forms an inexpensive and effective tool, which not only alters the ground reaction forces but also makes it difficult for the subject to adjust the motion and position of COM over the compliant (floating) BOS. When standing on the compliant surface, only the somatosensory information from the foot is distorted; in this case the other sensory inputs, namely the visual and vestibular, compensate for the loss.

When any two sensory components are diminished or lost, the effect is significant. Teasdale et al (1991) studied the effect of loss of vision and the somatosensory system on healthy young and older people. They found that the combined condition of foam and vision occlusion had a significant effect on balance and elimination of just one sensory system was not sufficient to differentiate between the older and younger population. Allum et al (2002) investigated the differences between trunk pitch and roll sway on twenty-five healthy young adults by incorporating a task of the SOT and a task on the sponge similar to those used in the CTSIB. Trunk sway was measured in standing position with and without vision on the following 3 conditions: (a) A foam support surface (b) Anterior-posterior (pitch) *sway-referenced* condition of the SOT, and (c) Standing on the same sway-referenced surface as in the second condition but with the body turned 90° so that sway was in the medio-lateral (roll) plane. The trunk sway was measured at the L-2 level using two angular velocity transducers, one for each plane. Peak to peak angular velocity and displacement were computed for pitch and roll planes. In the *pitch sway reference* condition there was a reduced trunk roll angle and velocity as compared to either the sponge or the roll sway conditions, however the trunk pitch angle and velocity were greater in this condition. For the *roll sway referencing* condition the trunk roll angle was greater than the other two conditions. For *the foam support* condition, roll velocity was highest. The foam condition also showed a multi-directional trunk sway. It was concluded that the use of the foam support surface was a more complex balance task than the sway referencing of the Sensory Organization Test. The use of foam induced sway in both pitch and roll directions, which was a limitation for the uni-axial sway referenced conditions of the SOT. This study advocated the use of the foam material as a more comprehensive balance assessing tool.

A critical issue seen in clinical practice is the challenges faced during outdoor walking. Even healthy people can find difficulty in maintaining or restoring balance after a disturbance on compliant or uneven surfaces.

This study will extend the concept of the sensory control of balance to investigate and to quantify the balance disturbances caused by uneven and compliant surfaces that people come across in outdoor walking. This will also provide some insight into the strategies used to recover from balance disturbances during walking.

2.3 THE MOTOR ASPECT OF BALANCE CONTROL

The two general balance control systems which operate to stabilize the body during normal stance as well as walking are feedforward and feedback control system.

Feedforward controls are pre-planned (preparatory) and predictive postural adjustments made in advance or simultaneously with movements to ensure that the stability is maintained (Dietz et al 1993). Some studies have used sinusoidal platform translations to study the feedforward mechanism. Dietz et al (1993) studied human stance and its feedforward control through a treadmill moving sinusoidally at different frequencies. Subjects stood upright on the treadmill with and without vision. The anterior turning point was defined as the time when the body changed direction from traveling forwards to backwards and the posterior turning point was vice-versa. EMG was measured for the right lower limb muscles, and the head acceleration was measured with respect to space. It was found that the adaptation to the changed frequency took place within four cycles; this was reflected in the change in the muscle activity as well as the kinematics. A feedforward mechanism was evident during the adaptation to the sinusoidal frequency, as after the four cycles there was a strong contraction in the tibialis anterior (TA) muscle during the anterior turning point of sinusoid and in the gastronemious (GA) muscle during the posterior turning point. This strategy helped in minimizing the movement by the body during the adaptation phase. There was no effect of vision on the changes observed. A similar study by Corna et al (1999) showed that in the absence of vision a large movement of the head was recorded with respect to malleolus, whereas, when vision was allowed, the body behaved in a manner of a single inverted pendulum. It was found that at the higher frequency of platform translation there was less coupling of body segments than at the lower frequency. This study showed that the balance control mechanism utilizes the information from vision in the feedforward control of stance posture. However, during a visual or any other sensory conflict situation, this mechanism can not prevent individuals from getting disturbed in an unexpected disturbance (Schieppati et al 2002). In such circumstances they require another mechanism which can restore their balance rapidly and prevent a fall.

Feedback controls occur after some unpredictable event requires a correction. This mechanism is responsible for the timely restoration of stability following a sudden and unexpected disturbance (Horak et al 1989). Corrective feedback responses have a rapid onset in the order of 60-100 ms onset latency and are whole body responses specific to the nature and direction of the disturbance (Pavol et al 2002). The time period for the feedback mechanism to operate and produce a corrective strategy is very short, and a failure of this timely correction can result in a fall.

The literature has defined three classes of strategies used in the feedback system that depend on the magnitude and speed of the disturbance and the size/type of the surface. With firm or large surfaces and small disturbances to the body, an ***ankle strategy*** takes place in which the whole body moves in one phase as a single inverted pendulum and only angular displacements about the ankle occur (Horak et al 1989). When the body encounters a fast or larger disturbance and/or is on a narrow/unstable surface, a ***hip strategy*** is recruited (Horak et al 1989, Szturm and Fallang 1998). Hips are the point of rotation with the head and hips moving in opposite directions to maintain the COM over the BOS. Here the body breaks up into a two-segment model, one being the upper body (Head-arms-trunk) and the lower legs. When the trunk bends forward there is an anti-phase motion of the legs i.e. ankle planter flexion. This so called hip strategy can result in rapid backward corrective movements of the COM relative to a stationary BOS (Szturm and Fallang 1998, Runge et al 1999, Fujisawa et al 2005, Park and Horak 2004). When a disturbance is too large, the ankle and hip in-place strategies are inadequate to restore the balance and a ***stepping strategy*** results (Maki and McIlroy 1996, Luchies et al 1994).

This strategy simply means quickly taking a step to move the BOS in order to meet the COM.

Szturm et al (1998) used different velocities and accelerations of platform translation and rotation to study the feedback strategy. It was found that the COM and the center of foot pressure (COP) parameters for toes-up rotation were different from those for the forward and backward translations. The onset of the displacement of COM, its peak magnitude and the time to reach the peak displacement scaled with the increasing speed for the platform rotation only. However, the magnitudes of other balance reactions like the peak displacement of joints and magnitude of muscle activity did vary as a component of speed in all the three conditions. Here multi-segment balance corrections were observed rather than pure ankle or hip strategy as seen by Horak (1986). The study also proved that the rotational disturbance was different from the translations. Wu (1998) compared the standing balance of the young and the older subjects by using a sudden platform translation. They found an increase in the movement of head and trunk for the older as compared to the younger subjects. Movement of these upper body segments for the older subjects showed an increased onset latency as well as time to reach peak amplitude of movement. However the thigh and shank segments for the elderly showed less amplitude of movement, decreased onset latency and shorter peak timings. These differences imply that older people use a different correction strategy to sudden platform perturbations which could render them more prone to instability.

A few studies have used multi-direction platform perturbations. Carpenter et al (1999) examined the balance reactions in young healthy adults while standing on a platform which randomly rotated in multiple directions. This study concluded that postural reactions are specific to the magnitude and direction of disturbances. The roll disturbances produce higher muscle activity and produce both roll and pitch balance corrections, whereas this is not true for pure pitch perturbations. This study also challenges the concept of pure ankle muscle response synergy because early para-spinal muscle activity was also noted. Maki et al (1996) studied the corrective stepping strategy in young healthy adults in response to different directions of platform translations. The objective of the study was to identify the characteristics of the swing leg during a non-sagittal perturbation. High, medium and low platform accelerations were used and subjects were instructed to take a step only when required. The results showed that for non- sagittal disturbances, stepping occurred for 91%, 32% and 2% for large, medium and small perturbations respectively and both right and left legs were used in equal proportion for stepping. For non-sagittal disturbances, most of the steps occurred with the leg that was unloaded in the perturbation. In this case, the step-length was also increased as compared to sagittal disturbances.

2.4 STUDIES ON NORMAL HUMAN WALKING

Walking is a more challenging task than standing because of the dynamics complexity associated with it. One gait cycle is from the heel strike of one leg to the next heel strike of the same leg. Normal walking has two well-defined phases (Perry 1992). The **stance phase** comprises a total of 60 percent of the gait cycle. This is the duration when the foot is in contact with the ground. It is further divided into: The heel strike, the mid-stance, the terminal stance and the pre-swing. The *heel strike* is the stage when the foot makes first contact with the surface. In this initial 10 percent of the gait cycle the foot starts to load. This is the stage of double support when both feet are on the ground. This is considered to be the most stable duration during the gait cycle. The *mid-stance*, occupies 10-30 percent of the gait cycle. Here the trunk continues to progress forwards and the body is in the single support duration. The *terminal stance* occupies the next 30-50 percent of the gait cycle. Here the heel of the support foot starts to rise and that ends the single support duration. The *pre-swing* is the last phase of the stance, occupying 50-60 percent of the gait cycle which begins with the heel contact of the swing leg and ends with the toe-off of the stance leg. The **swing phase** constitutes the next 40 percent of the gait cycle and has following main stages: the initial swing, the mid-swing and the terminal swing. During the *initial swing*, foot clearance is the main objective. In the *mid-swing*, the limb further advances and crosses the opposite limb. The *terminal swing* is marked with the end of limb progression and concludes when the foot strikes the ground. The adaptability to meet changes in the environment or walking surfaces makes gait a difficult task. In the steady state of walking, which is attained after the second step (Jian et al 1993), there are times during the single support phase where the whole body weight is on the stance leg. Further, the large distance of the COM from the constantly moving BOS and the long duration of the single support period make walking a complex task (Winter et al 1993). Analyzing balance reactions during gait thus needs special attention. Studies have described human walking by analyzing the kinematics including the linear or angular positions and the velocities or acceleration of body segments.

A number of studies have analyzed leg muscle EMG signals during walking. During normal walking, TA and extensor digitorum (ED) show highest activity at the heel-strike and before the toe-off. GA and soleus (SL) have maximum activity during the mid-stance and the terminal stance respectively. Peroneus longus (PL) and peroneous brevis (PB) have a major burst of activity during the heel off phase (Hunt 2001). The knee and hip extensors support the body at the initial stance phase. The lengthening of Rectus femoris (RF) in the later part of stance reduces the speed of the leg and disperses power to the trunk for its forward movement. The swing phase is initiated with the passive stretching of hip structures, causing the hip flexors to contract and is terminated by the lengthening of hamstrings which decelerates the movement of the leg (Neptune et al 2004). Otter et al (2004) further explored the EMG response of lower limb muscles during a systematic decrease in the walking speeds. In general, the amplitude of the muscle activity reduced with the decrease in the walking speeds. However, as the speed went to 0.28m/s and lower, an increase in the activity of PL during the single support phase was observed. The authors felt that this could be attributed to an increased demand for the stabilization of the stance leg and prevent excessive foot inversion. The increase in the activity of biceps femoris (BF) at the late stance and the early swing could be for the facilitation of hip extension and knee flexion. RF activity was increased in the late swing and could be due to the extra force required to generate hip flexion because of the decrease of momentum at low speeds. This study provides some insight into the fact that during very slow walking there is an increased demand on neuromuscular control during the single support phase and a need to generate more force in the clearance of the swing leg.

The clinical significance of studying balance while walking is even more important with respect to the older population, as around 70% of falls in the elderly occur during walking (Cali et al 1995, Berg et al 1997). The declining effect of the balance control system becomes more evident with the older population as a result of the physiological changes accompanying aging. With increasing age, there is a decline in various parameters related to the maintenance of mobility such as the musculoskeletal system (muscle strength, flexibility and joint range of motion), the vestibular and the visual system. In the older population it becomes difficult to control balance during the dynamic tasks such as walking and usually the risk of fall increases during such activities (Hagemon et al 1986). This has led researchers to investigate the age-related changes which affect mobility and walking performance (Gabell et al 1984; Winter 1990).

One of the earlier studies conducted to analyze the gait pattern of the older versus the younger population was by Gabell et al (1984). Their study showed no significant difference in stride width and double support duration between the groups. On this basis the researchers found that the increased risk of falling during walking is not because of the age factor. However, the only spatio-temporal variables they analyzed were stride width and double support duration. The study did not control for confounding variables such as the speed of walking. Winter (1990) conducted a similar study to compare kinetic (force related parameters) and kinematic (motion related variables) gait variables between healthy young and older populations. Subjects walked at a self-selected pace on a leveled walkway. Major differences found between the two groups were increased stance, double support duration, and a reduction in the stride length for the older subjects. Covariance of hip and knee moments was taken as an index of coordination. There was a significant reduction in the covariance of hip and knee moments of force in the older individuals. From this measure the researchers inferred that the older individuals were unable to make an anterior posterior shift in the pattern of movement to control their balance in the saggital-plane. Kerrigan et al (1998) studied differences in the peak joint movement, moment and power between healthy elderly and young adults at self-selected and fast walking speeds. At the self-selected walking speed there was a significant difference in 11 out of 28 kinematic/kinetic variables between the older and younger individuals, but at the fast speed most of the differences did not persist between the two groups and only 4 parameters showed significant group difference. These were: reduced hip extension, high anterior pelvic tilt, decreased planter flexion and decreased ankle power generation for the older subjects. Similar findings were observed by Riley et al (2001). They found that at similar speeds, hip and knee moments of the older and younger subjects were comparable.

These two studies offer some insight into the fact that most of the kinematic and kinetic differences between the younger and older groups can be significantly attributed to the difference in walking speeds. Shkuratova et al (2004) compared the walking strategies in young and old subjects during the following 4 tasks: walking in a straight line at self-selected and fast speeds, walking in a 'figure of eight' at self-selected speed and walking in 'figure of eight' with a secondary task of transferring coins from one pocket to another. During the task of walking in a straight line at fast speed, the elderly showed a significantly lower speed, shorter stride length and increased double support duration as compared to the younger subjects. With the turning task (walking in a 'figure of eight'), the walking speed was reduced in both the groups as compared to straight line walking. No significant group difference was observed in the coin transfer task, thus increase in task demand did not alter walking strategies in the two groups.

2.5 STUDIES RELATED TO PERTURBATION DURING WALKING

The above studies have focused on self-paced walking on a fixed firm surface. Although these studies provide important background information about walking parameters, they do not evaluate feedback balance control mechanisms during walking. Only a few studies have focused on balance challenges during gait due to different surfaces or environmental conditions. Mechanically perturbed human gait is not often examined as it is difficult to produce a systematic balance disturbance and to control its timing and magnitude. Some of the common methods applied to study disturbance during walking use the obstacle paradigm (Eng et al 1994, Chou et al 2001), platform translation (Tang et al 1998, Riediger et al 1999) and uneven or compliant surfaces (Menz et al 2003).

Eng et al (1994) studied the response to an obstacle hitting the left leg during the early and the late swing in over-ground walking. The positioning of the obstacle was triggered when the right foot contacted the first force plate. The left foot momentarily came to a stop after hitting the obstacle. It was found that the perturbation during the early swing phase resulted in an elevation strategy and a flexor pattern for the swing leg, whereas the late swing perturbation resulted in a lowering strategy and an extensor pattern. Balance corrective response after a trip was phase dependent. A similar study was done on a treadmill by Schillings et al (1996) to study stumbling reactions in young adults. An obstacle controlled via electromagnet was dropped to disrupt the early swing of the left leg. The results were similar to the study by Eng et al (1994). The advantage in using treadmill is that it controls the walking speed and timing of disturbance however it forces the action of the legs and there is no forward propulsive movement by the subject.

The moving platform paradigm is another approach to study the feedback aspect of balance while walking. Tang et al (1998) carried out a study on healthy subjects at a steady state of over-ground walking. Platform translation was triggered with the heel strike of the right foot. Reactive response was noted consistently in bilateral leg and thigh muscles namely Tibialis Anterior (TA), Rectus Femoris (RF) and Biceps Femoris (BF). The magnitude of response of the proximal muscles, particularly the abdominals and the gluteus medius was attenuated with repetitive trials. Activity in the contra-lateral TA, RF and BF was also noted as a response to the perturbation. This study was based on the work by Nashner (1980) and Dietz (1985) that showed that the ankle muscle activity resulting from the perturbation is the main input for balance correction during walking and ankle motion are sufficient to restore balance during perturbation on a treadmill. However, it is well established now that the response to a larger disturbance is not restricted to the ankle joint only but can be multi-segmental as seen in the hip or stepping strategy (Reidiger et al 1999; Szturm et al (1998).

Reidiger et al (1999) examined the ability of the postural control mechanism in restoring balance after perturbations during two different phases of gait. Healthy young and older individuals walked on a platform which translated in an antero-posterior direction. The platform translation was triggered with the heel off of the right leg at two stages corresponding with the initial and the mid swing respectively (Perturbation 1 and 2). The balance reaction was taken into account after the onset of the earliest muscle activity. It was found that there was no significant difference in the corrective balance strategies and in the magnitude of kinematic variable for the two translation conditions except that there was a phase shift of the trajectory of the kinematic variables between the two perturbations. This showed that the balance adjustments and walking task are closely integrated rather than two separate processes. The corrective balance response was also noted at ankle, knee and hip in both perturbation conditions; which showed that there was a multi-segment response to the perturbation rather than the pure ankle muscle response as proposed by Nashner (1980), Dietz (1985) or Tang et al (1998).

Oddsson et al (2004) carried out a similar study in which they analyzed the behavior of the body after a disturbance during the stance phase of the gait. Subjects were asked to walk without shoes on a level walkway. Infrared markers were placed at the sternum and the shank of both legs. Two types of diagonally directed platform translations were introduced around 200 ms after heel strike at a steady state walking. They took the medio-lateral displacement of the sternum marker as a measure of the COM. However, this may not be a reliable measurement of the COM because it has been shown that the trunk can have large amplitude of movement with leg motion in opposite direction as in the hip strategy, this would have little effect on the position of COM (Szturm et al 1998). The study found that the medio-lateral stance width scales in magnitude with an increase in the intensity of the disturbance, and the body behaved as a single inverted pendulum. They did not specify whether this distance was measured for the passive or an active component of the movement. This is an important distinction because the passive component of the medio-lateral distance does not reflect the corrective strategy and will definitely increase or decrease with the type of disturbance and the direction of platform movement.

Most recently, studies have used compliant or uneven surfaces to challenge walking. This is another approach to look at the feedback aspect of balance control in a more functional way. In a series of studies, Menz and colleagues (2003a, b, c and d) studied three different groups of subjects including young (age-group 20-40), elderly (age-group 70-90) and diabetics to analyze acceleration patterns of the head and pelvis while walking on a regular and an irregular surface. The 20m long irregular surface consisted of a continuous sheet of 2cm thick foam and several 2cm thick wooden blocks of different shape and sizes placed randomly underneath a 5mm artificial grass carpet. The experimental surface thus made, potentially introduced an uncertainty during the walking task. However, it incorporated an error for the consistency of the task as the wooden blocks were randomly distributed. The study did not mention if all the subjects stepped on the wooden blocks or not. The dependent variables calculated were walking velocity, cadence, average step length, step timing variability, acceleration amplitude variability and acceleration for the head and pelvis. Younger subjects did not alter their walking velocity on the experimental surface but they showed a reduction of cadence, an increase in the step length and an increase in the step timing variability on the experimental surface. They found that on the irregular surface, the pelvis showed significantly greater acceleration amplitude variability in all the three axes, whereas there was no significant difference in the head acceleration. It was concluded that maintenance of the head stability was the main task of the postural system while walking on an irregular surface (Menz et al (2003a). In the study with the older subjects, Menz et al (2003b) showed that the older subjects had slower walking speed, shorter step length, greater step timing variability and a reduced acceleration RMS at both head and pelvis for both the surfaces as compared to their younger counterparts. Elderly subjects with high risk of falling had a lower magnitude and more variable pattern of acceleration at both the head and pelvis as compared to the elderly with low-risk of falling (Menz et al 2003c). These results can be confounded with the fact that the older subjects were walking at a slower speed. Studies have shown that several kinematic and kinetic differences between the young and elderly did not persist when walking speed for older subjects was increased and at similar speed hip and knee moments for young and old were comparable (Kerrigan et al 1998, Riley et al 2001). Changing the gait speed entirely changes the kinematic and kinetic variables and thus can not be considered as the base line for comparing other gait variables for the two age groups (Lelas et al 2003).

Recently researchers have started analyzing changes in the walking pattern of subjects having diabetic peripheral neuropathy (DPN). These patients form an interesting group to study balance while walking. They have reduced plantar sensation, intrinsic foot muscle atrophy and decreased muscle strength that account for impaired balance and falls (Gutierrez et al 2001, Eils et al 2004). Menz et al (2004d) carried out another descriptive study to examine and compare acceleration amplitude variability of the head and pelvis of older adults with diabetic peripheral neuropathy and their age matched controls, diabetics without any neuropathy. The experiment protocol was the same as described above in the other studies of Menz and colleagues. They found that the older subjects with DPN had slower walking speeds, smaller step lengths and increased step timing variability on both level and uneven surfaces. The acceleration pattern showed that signals for the DPN group was less in magnitude but more variable at the head and pelvis than the control group. From this the researchers concluded that subjects with DPN had less ability to stabilize their head despite their slower walking speed. However, an important point to consider in these experiments is that the task for all subjects on the experimental surface can not be considered the same. The results do not show whether subjects stepped on the wooden blocks or not. Acceleration results could be different because of the difference in the step length due to different speeds of the subjects. The consistency of the task and the control of speed thus becomes an important consideration in studies incorporating over-ground walking.

2.6 FOOT MECHANICS AND PLANTAR PRESSURE DURING WALKING

The foot makes contact with the ground or support surface and forms a very important link for sensing the surface characteristics and for propulsion and stability during walking. Neuromuscular changes involving the foot segment, such as the absence of plantar sensation or muscle weakness can not only make a huge difference in the loading pattern during walking but can also change other joint kinematics (Eils et al 2004). A number of detailed motion analysis studies have described the foot segment as a multi-segment model during walking (Kidder et al 1996, Leardini et al 1999). The divisions were as follows: (1) *The Hind foot-* calcaneous, talus (2) *The Mid-foot-* navicular, cuneiforms and cuboid *(3)The Fore foot-* metatarsals and (4)*The Hallux.* The data for the original position of markers were taken in the stance position of subjects and then compared with the data of subjects walking at a self-selected pace. The movement of the segments was analyzed in saggital, coronal and horizontal planes with respect to its proximal part. During the **heel strike**, the hind-foot goes into plantar flexion, the mid-foot into supination, the fore-foot into dorsiflexion, adduction, pronation and the hallux in a relative dorsiflexion and internal rotation. During the **mid-stance** the hind-foot shows a relative dorsiflexion, the fore-foot goes into valgus and the hallux is in plantar flexion. At the **toe-off**, the hind-foot goes into plantar flexion, the forefoot goes in plantar flexion and supination and the hallux moves into extension. The **Swing phase** begins with the hind-foot moving into plantar flexion, the fore-foot into dorsiflexion and abduction and the hallux remains in a valgus position through-out the swing phase.

Foot pressure is an important parameter in the evaluation of foot function. In the analysis of the foot during walking, foot contact and its interaction with walking surfaces is very important. It is usually measured by some pressure mapping system, the most common being the *force platform* used in posture and gait analysis. These calculate the resulting ground reaction forces, free moment and also furnish displacement of the instantaneous COP. The measurement of COP excursion provides accurate information about postural control (Geurts 1993). Force platforms have been found to deliver reliable information about the resulting vertical reaction. However, they do not provide information about the local distribution of pressure under the foot, and probably more importantly restrict analysis to the firm fixed surface of the force plate. Pressure or load distribution and COP on different surfaces can be determined using pressure mapping systems. *Pressure mats* consist of a an array of pressure sensors, which not only tell us about the pressure distribution under different areas of the foot but also determine the foot location and the trajectory of the COP. The only thing which is neglected by these pressure mapping systems is the shear stress and vertical reactions. A novel method to analyze the foot contact was devised by Giacomozzi et al (1997). This research group incorporated both the pressure platform and the force platform and named the system as the piezo-dynamometric platform for a more comprehensive analysis of the foot pressures. This new device was formed by tightly securing the pressure platform on top of a force platform. The data was acquired simultaneously from both devices and a software corrected the redundancy and integrated the data. This new instrument incorporated both force and pressure platform data and seems to be useful in a comprehensive analysis of floor to foot interaction.

Warren et al (2004) put forward a detailed analysis of plantar pressure for nine different regions of the foot and correlated the results with different gait speeds ranging from 0.45 m/s to 1.79 m/s. Subjects included healthy men who wore an EMED Pedar in-shoe plantar pressure measurement system (Novel electronics) and walked on a treadmill at different randomly assigned speeds. The data collection started at every heel strike of the right foot and was taken till the 10^{th} step of the right leg. The general gait pattern observed at all speeds was a typical heel-toe gait pattern where plantar pressure in the heel region increased followed by an increase in the mid-foot and thereafter in the forefoot region. There was no change in the general shape of the curve, however as the speed increased, pressure in the forefoot and toe region peak slightly earlier in the gait cycle. For example, the peak pressure in the forefoot region occurred at 52% of the gait cycle at the lowest speed and at 45% of the gait cycle at the highest speed. In the toe region it occurred at 59% at low speed and at 53% at the highest speed. With an increase in the speed, peak values of plantar pressure increased linearly except in the central forefoot and lateral forefoot region. One reason for this can be the fact that the central forefoot showed greatest peak pressure at all speeds. The greatest increase in the peak pressure was noted in the toe region which rose by 289%. The heel region showed the next greatest increase in the plantar pressure (124%). Medial forefoot showed a 91% increase and the central forefoot peak pressure had a 49% increase till the speed of 1.34 m/s and leveled thereafter. This study provides important information regarding the variation of plantar pressure with respect to the walking speed. Thus, in an experiment involving plantar pressure analysis during human locomotion on regular or irregular surfaces, the control of speed would be an important parameter to consider. The study was done on a treadmill which could be a different factor from the subjects walking on the ground. Another limitation of this research is that it has not taken into account the relationship between the times spent in each region of the foot and the amount in pressure rise.

An attempt to look at the plantar pressure with respect to foot sensation was carried out by Nurse et al (2001). In this cross sectional study they examined the effect of altered foot sensation on the plantar pressure during walking. The sole of the left foot was desensitized by application of ice prior to the trial. Three different zones of foot were selected for desensitization: *The whole foot, the forefoot,* and *the rear foot.* Young healthy subjects recruited for study had reported the absence of any neurological dysfunction or disease. The left foot was divided into following 7 areas: heel, medial arch, lateral arch, 1st metatarsal head, metatarsal heads, hallux and toes. For each area, the path of COP, the maximum plantar pressure (P_{max}) and the pressure-time integral (P_{int}) was measured using flexible insoles, which were attached to the plantar aspect of the foot. Subjects were asked to walk bare foot on a level surface indoors. The results showed that there was a shift in the maximum plantar pressure from areas of reduced sensation to areas of normal sensation. For example, in the case of *rear foot* cooling the P_{max} and P_{int} were reduced at the heel and in the case of *fore foot* cooling the P_{max} and P_{int} were reduced at the toes. For the *whole foot* condition, these two variables were reduced at the heel as well as toes but were increased at the metatarsals heads. The COP trajectory also showed a shift in response to cooling. There was an anterior shift of the COP trace in the rear foot sensory loss and a posterior shift in the forefoot sensory loss.

A number of performance based scales have been developed to assess balance, For example, the Functional Reach Test (Duncan et al 1990), the Timed Up and Go (Podsiadlo and Richardson 1991) and the Berg Balance Scale (Berg et al 1995).

Some tests involve multiple tasks such as the Tinniti's performance oriented assessment of mobility (Tinniti 1986). This test deals with objective evaluation of balance while performing tasks in standing and walking. However, Tinniti's test has poor sensitivity and also fails to find subtle changes in mild balance problems (Horak 1997). The Berg balance scale consists of 14 items and the tasks are only for the assessment of standing balance. These tests do not include unpredictable effect of different surfaces. Many researchers have analyzed strategies used to recover from sudden brief mechanical perturbations during standing. Inducing sudden single platform translation (Szturm et al 1998) and use of compliant surfaces (Menz et al 2002) has been the most popular paradigm used to study the feedback mechanism. Research studies have focused more on balance while standing in place with a stationery base of support and fewer studies have been carried out on balance while walking.

2.7 SUMMARY

There is a need to develop a walking function protocol for objective outcome measures, which will delineate the subjects having a balance problem during walking ability (Shimada et al 2003). Attempts have been made to study feedback strategy during walking using expensive and elaborative laboratory equipment, such as a movable walkway platform (Tang et al 1998, Riediger et al 1999). Balance requirements during walking are not restricted to indoor walking on level surfaces only. An important consideration for walking performance is the ability to negotiate outdoor terrains which can be unpredictable. Different walking surfaces and terrains may produce uncertainty and an increased threat to balance control. Use of sponge pads and other similar compliant surfaces are becoming popular in studying balance during walking (Menz et al 2004). However, no study so far has provided a balance disturbance using compliant surfaces in a systematic manner.

The present study has analyzed balance while walking on regular and irregular or compliant surfaces. The balance strategies have been quantified from the correlation of the accelerometer signals for the upper trunk and shank, the COM computed from VICON video based motion analysis, gait parameters evaluated from the GAITRite® carpet and the plantar pressures from the pressure mats (FSA – Vista Medicals Limited). Balance control in both sagittal and frontal planes has been evaluated.

3. PURPOSE, OBJECTIVES AND HYPOTHESIS:

3.1 PURPOSE

The purpose of this study is to develop a task protocol for the evaluation of the foot to surface control, and balance requirements of steady-state walking on compliant and irregular surfaces that would represent typical out-doors terrains.

3.2 OBJECTIVES

1. To determine whether the task protocol was comparable for all the surfaces and there was no anticipation for the type of surface to be encountered. Any deviation in the gait parameter before the surface will account for the power of anticipation.

2. To identify the type and magnitude of balance disturbance caused by stepping over different surfaces representing outdoors. This can help to develop a method, to create a series of graded disturbances while walking. Changes in the spatiotemporal gait parameters during stepping onto different compliant and irregular surfaces would reflect the magnitude of the disturbance.

3. To identify and measure the balance reactions and motor strategies used by healthy subjects to compensate for the mechanical disturbances produced by altered support surfaces during steady state walking. The changes in the gait parameters for steps after the surface, deviation in the body centre of mass and the plantar pressure will be analyzed.

4. To examine motor coordination while correcting for stumbling disturbances produced by altered support surfaces. Trunk and shank accelerometer signals will be cross correlated to determine the index of movement coordination between the upper body and lower limbs.

3.3 HYPOTHESIS

During walking, stepping onto an irregular and/or compliant surface will cause mechanical disturbances which will be corrected at the level of foot and ankle and will not result in significant change in the whole body centre of mass and the coordination between body segments.

4. METHODOLOGY

4.1 SUBJECTS

Twenty subjects, aged 20-40 were recruited for the study from students and staff of The University of Manitoba. Subjects were screened by oral questionnaire, to rule-out any history of neurological, musculo-skeletal, visual or vestibular problems. They were fully informed about the experiment protocol. The University of Manitoba Ethics Committee approval and written consent from the subjects were obtained prior to their recruitment in the study.

4.2 EXPERIMENTAL PROTOCOL

The subject's height, weight and activity level was measured before the onset of experiment. Subjects wore shoes without heels, t-shirt and shorts, to allow easy placement of accelerometers and VICON markers on body segments. They wore the pressure mat in the shape of a shoe-insert on the right side for recording parameters related to plantar pressure. Subjects were asked to stand still before each trial. When instructed they started walking at their preferred pace for a distance of five meters and then stopped. The data capture area was the middle three meters.

The following surfaces were used in the study.

1. *Control surface*: A firm insulating material of one inch thickness and 2.5 m long covered with a thin-pile carpet.
2. *Foam grid surface (1cm):* This surface was 1m in length. It had medium density sponge cubes spread at a distance of 1cm forming a uniform grid.
3. *Foam grid (4cm):* This surface was 1m in length. It had medium density sponge cubes spread at a distance of 4cm forming uniform grid.
4. *Dowels surface*: This surface was 1m in length. Dowels of 3cm diameter were spread (on their sides) at a uniform distance of 6cm.
5. *Compliant surface*: This was a 3m long exercise mat of 3cm height.
6. *Ridges surface:* This surface was 2.5m long. 3cm deep ridges were 6cm apart.

During a trial, one of the experimental surfaces was placed underneath the carpet in the middle three meters of the vicon capture area. In total forty walking trials were collected with four trials on each of the five experimental surfaces and the rest on the control surface. The order of the six surface conditions (one control and five experimental) was randomized. The surface was arranged in such a way that the subjects stepped on the surface with their right foot. Rest was provided if the subject requested, or appeared to be fatigued. A physiotherapist was standing beside the subject to provide assistance if needed. A trial in which subjects needed support was considered as a fall.

4.3 MEASURING INSTRUMENTS

An electronic walkway GAITRite® was used to capture temporal and spatial gait parameters. GAITRite® can be compared to a portable carpet having a length of 525cm with an active area 366cm long and 61 cm wide. Data was first collected by on-board processors attached to the mat which are then uploaded to a computer. The GAITRite® software processes the data and outputs the means of all the spatial and temporal gait parameters. McDonough et al (2000) established the concurrent validity of the GAITRite® carpet with paper and chalk method for spatial parameters and with a hand-held stop watch for temporal variables. The validity of GAITRite® has also been established against video based motion analysis system (Cutlip et al 2000) and standard foot switches (Bilney et al 2003). Its test-retest reliability has been established for separate measurements taken over one day Bilney et al (2003) and one week (Van Uden 2004).

Miniaturized tri-axial accelerometers (Nextgen, is 2x1x1 cm in dimension and 30 grams in weight), which can be secured to the skin, were used to record segment motion. Nilssen (1998) tested the reliability of trunk accelerometry for the task of standing and walking on level and uneven ground. In this study the accelerometer was placed in the lower lumbar region. However this location does not provide a proper description of how the lower extremities and the upper body behave and contribute to the displacement and velocity of the COM and/or COP. The result of the above study reported a high reliability of trunk accelerometry for both standing and walking on the level surface. On the uneven surface the reliability was high except for the medio-lateral axis. In the present study accelerometers were placed on the T2 vertebra (trunk segment), and on the lateral malleolus of the left leg. The accelerometer signals were recorded and stored on a computer and were analyzed using the Matlab 6.5 software.

The FSA has a piezo-resistive semi-conductive polymer between two layers of highly conductive nylon fabric. This whole structure is encased in a protective cover of polyurethane. The final cover is made up of a stretchy Lycra cover. The FSA mat used for this experiment was in the form of an insole and was connected to the interface module through a serial interface cable. FSA 3.1.x version software was used to read the signals. The thin flexible shoe-insert, placed inside the shoe, consisted of an array of 128 miniature pressure sensors of one-quarter square inch. The pressure sensors were calibrated in absolute pressure in 8-bit resolution, and the sampling frequency used was 35 Hz. The COP in medio-lateral and anterior-posterior directions was calculated from the recorded insole pressure values.

The VICON 460 video motion analysis system was used for whole body motion analysis (Jones, et al 2005, Lamontagne 2005). Six digital video cameras recorded 3D coordinates of markers placed over various body parts (refer to figure 1 for the marker positions). The 3D coordinates of the centre of mass were computed, using Vicon's Plug-in Gait Software.

A custom trigger electronic circuit was used to connect the FSA and Vicon systems to synchronize data collection of the insole pressure sensors of the FSA data recording software and the Vicon data recording software (both the digital marker coordinates and the analog-to digital converter used for accelerometer signals.

4.4 DATA ANALYSIS

The dependent variables in the study were as follows:

- *Spatiotemporal gait parameters*: Stance time, swing time, single–double support durations, step length, step width and stride velocity was calculated for the two steps before the surface and the three steps after the surface. Stride velocity is calculated from heel strike of one leg to the heel strike of the same leg. Any variation in these parameters before the surface would show that the subjects were anticipating the type of surface to be encountered and variations after the surface would show the effect of the irregular and/or compliant surface on gait.
- *Peak plantar pressure* and *pressure time integrals* in the five different zones of the foot – medial heel, lateral heel, mid foot, medial forefoot and lateral forefoot. Pressure time integral (PTI) is defined as the amount of loading in a particular zone of the foot when the foot comes in contact with the ground. It was calculated as the product of pressure applied in a particular zone of the foot and the time during which the pressure was applied. PTI not only reflect the total pressure but also the duration and studies have shown that it is a better indicator of foot function than the total pressure (Chen et al 1995).

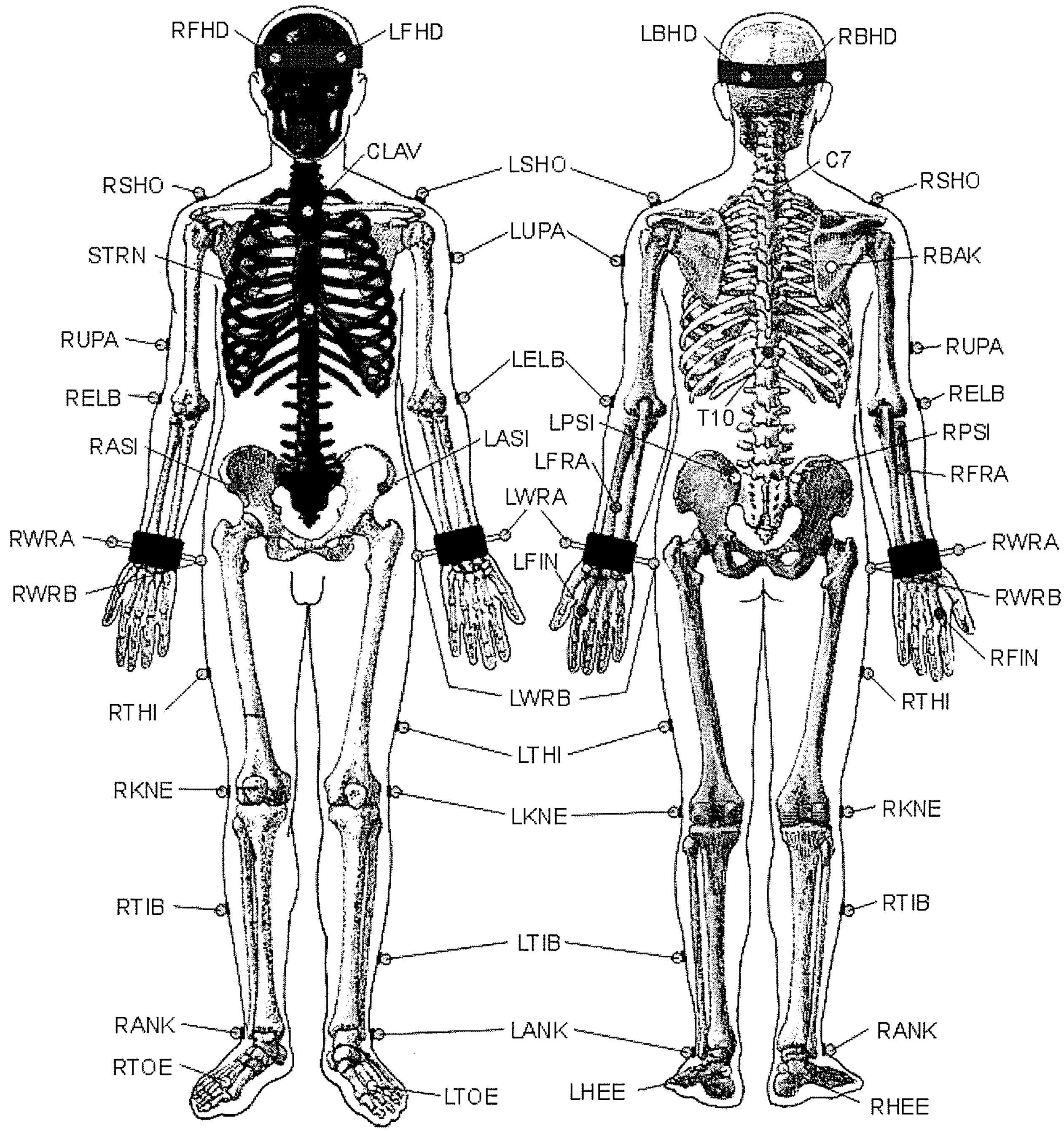

Figure 1 – Illustration of the full-body marker positions for the VICON Video motion Analysis (Vicon Motion Systems)

- *Maximum excursion of COM* in medio-lateral direction for the first step on the surface was examined. This was computed using Vicon's Plug-in Gait Software. Studies have shown that the COM excursion can be an important descriptive of variation in gait and is used by clinicians to assess walking problems (Iida et al 1987; Engsberg et al 1992). It has also been used to assess gait efficiency and symmetry (Tesio et al 1991).

- *Maximum excursion of COP* in medio-lateral and anterior-posterior direction was determined. Studies have shown that the displacement of COP is an indicator of postural stability and increase in the COP displacement is directly related to amount of muscle activity during a disturbance (Nakamura et al 2001).

- *Index of stability*: The cross-correlation function (peak r value and phase) between COM and COP was determined as an index of stability. For regaining balance after a disturbance, a body should have enough momentum to position the COM over the base of support (Pai and Patton 1997). This value shows how stable the subjects were while walking on the experimental surfaces. A second index of stability was defined as the peak 'r' values between COM and the ankle angle.

- *Index of movement coordination:* The cross-correlation function *between trunk and leg acceleration signals* was defined as the index of movement coordination. The correlation function (peak r values) between trunk and leg acceleration signals was determined as an index of how well these body parts are coordinated as a function of walking surface. This measure shows the impact of the disturbance on the coordinated movement.

The Cross-correlation function is used to measure the relationship between two signals. When the two signals are similar in nature and overlap in time with respect to each other, their product is like a constructive interference, and the result is a positive value. When two signals are similar but are in opposite direction, the 'r' value is negative. The magnitude of this 'r' varies between ± 1. Values closer to 1 indicate stronger relationship between the two variables. For example, in an inverted pendulum like in the ankle strategy model where the COP can define the COM and vice versa, cross correlation gave a high positive value. But in a situation where there is hip strategy or multi-segmental motion, the values can be negative and or can show a lead or lag of one segment over the other. The following studies have used cross correlation to look at the relationship between movements of various body segments and analyzed it as a measure of balance coordination.

Mesure et al (1997) utilized the cross-correlation function of head and hip lateral accelerations to examine the effect of sports training on postural disturbances during standing on firm and sponge surfaces. Here the peak of the cross-correlation function was signified as a measure of head-hip coordination. The Cross-correlation functions from each trial were averaged and the peaks of the amplitude of common significant peaks were subjected to ANOVA. The peak amplitude, which is the intensity of correlation between the head and hip, showed significant differences between the hard and soft surface.They concluded that sports training results in improved segmental coordination in healthy subjects.

Gatev et al (1999) studied carried out cross-correlation analysis to analyze the behaviour of various body segments during standing with normal BOS or with a narrower BOS and with and without vision. The body segment motion and points that were cross correlated were- COP and COG, COP and ankle angle, COP and knee angle, COP and hip angle, A-P and M-L knee and shoulder motion, and head and COG motion. With the narrower support surface, the peak correlation value between the COP and COG decreased showing that subjects become less stable with a narrower base of support however, there was little change with vision

Nardone et al (2000) cross-correlated malleolus, hip and head segment of elderly subjects standing on a sinusoidally moving platform. They found that with the increase in age and at higher frequency of platform oscillations, there was less coupling of head and hip segments.

5. STATISTICAL ANALYSIS

The **independent variable** used in this study was *surface* which had six levels. One c*ontrol surface* and five experimental surfaces – *Foam grid* (1cm), *foam grid* (4cm), *dowels surface*, *exercise mat* and the *ridges surfaces*. The **dependent variables** used were 1) Spatiotemporal gait parameters; 2) Peak plantar pressure and pressure time integrals; 3) Maximum excursion of COM; 4) Maximum excursion of COP; 5) Index of stability; 6) Index of movement coordination.

Initially, before analyzing the effect of the surface conditions on the dependent variables, it had to be determined if the walking task was comparable for all the surface conditions. This was determined by the comparison of the gait variables between surfaces for the first two steps (These were the steps before the subjects encountered any type of surface). This was done in order to determine whether subjects anticipated the type of surface. The dependent variables for the steps after the surface were analyzed to determine the effect of the surface condition. Each step was analyzed separately rather than averaging the variables for the steps.

The univariate analysis of the General linear model (ANOVA) was carried out using the SPSS statistical package (version 13) for all the comparisons. An alpha of 0.05 was used.

7. RESULTS

The main objective of this study was to identify and measure the balance reactions and motor strategies used by healthy subjects to compensate for the altered support surface conditions during steady state walking.

6.1 Group comparison for the consistency of the walking task

The walking task consisted of 6 steps. The participants started walking with their right leg and they always encountered the surface with the 3^{rd} step (right – left – right). This was done to ensure that the participants had achieved the steady state of walking before they encountered the surface. An analysis of the effects of surface condition on spatial-temporal gait parameters of the two steps preceding the modified surface was performed to determine whether the walking task was different for any of the experimental conditions. **Table 1** presents the group means, standard error of means for each of the six surface conditions and the F values resulting form the univariate statistical analysis. It shows that there was **no statistical difference** observed in the **first two steps** for the stance duration, step length and step width between the six surface conditions. This demonstrates that the walking tasks were performed in a similar fashion, independent of the support surface type encountered on the third step, and consequently did not change the way subjects approached the potentially disturbing surfaces.

Table 1 – *Result of the univariate statistical analysis for spatial-temporal Gait parameters: Mean and standard error values for the six surfaces and the Level of Significance (F value)*

	First Step (Right)		**Second Step** (Left)	
Parameter	Surface Means (S.E)	Level of Significance (F Value)	Surface Means (S.E)	Level of Significance (F Value)
Stance Duration (sec)	1- 0.762 (.015) 2 - 0.711 (.015) 3 - 0.737 (.015) 4 - 0.709 (.015) 5 - 0.709 (.015) 6 - 0.707 (.015)	**0.07** (2.162)	1- 0.660 (.006) 2 - 0.656 (.006) 3 - 0.663 (.006) 4 - 0.664 (.006) 5 - 0.667 (.006) 6 - 0.665 (.006)	**0.9** (0.359)
Step Length (cm)	N.A.	N.A.	1- 67.211 (.988) 2 - 69.939 (.988) 3 - 71.257 (.988) 4 - 70.252 (.988) 5 - 69.197 (.988) 6 - 70.292 (.988)	**0.09** (1.963)
Step Width (cm)	N.A.	N.A.	1- 69.353 (.902) 2 - 71.563 (.902) 3 - 72.762 (.902) 4 - 72.135 (.902) 5 - 71.183 (.902) 6 - 71.927(.902)	**0.1** (1.693)

(* Significant difference)

6.2 General Findings

During the experiment protocol none of the subjects fell or stopped walking. All subjects stepped on the surface with the third step of their walk (right leg). In total there were 20 dependent variables which included 6 temporal gait parameters, 2 spatial gait parameters 10 plantar pressure parameters (including pressure time integral and peak pressures for the five different zones of foot), 1 index of movement coordination (correlation coefficient between trunk and leg accelerations) and 1 index of stability (correlation coefficient between ankle angle and COM).

The dowels surface had the maximum number of parameters that were significantly different from the control surface (16/20). The surface with ridges and the foam grid (1cm), ranked second on which 15 of 20 dependent variables showed a significant difference. The foam grid (4cm) showed 13 of the total variables as significantly different and the exercise mat had 10 variables significantly different from the control surface.

6.2.1 Spatial-temporal gait variables

Table 2a describes the Mean values for the spatiotemporal gait parameters for the third, fourth and the fifth step on surfaces 1 to 6 and table 2b shows the results of ANOVA for **the main effects of surfaces** on spatial-temporal gait parameters. With the exception of the step length and width of first step on the surface and the stance duration of the next left step and the all the variables showed highly significant differences on the experimental surfaces.

Figure 2 presents the group means and standard error of means for the spatial-temporal gait parameters for the third step of the walk, which was the right foot's first contact onto the different surface conditions. The top two histograms show the stance and the swing durations. The middle two present the step length and the step width and the bottom two presents the single support and the double support durations. Horizontal axes have six bars representing the six different surfaces. The post hoc analysis showed the following results. For the **first step on the surface**, the compliant surface and the surface with ridges had significantly **lower** values for the stance duration as compared with the control surface. The average difference in the value was 50 ms ($p<0.0001$). For the swing duration, with the exception of the compliant sponge surface all other surfaces showed significantly **lower** values from the control surface with the average difference of 35ms ($p<0.0001$). Step length and step width did not show any significant difference between the groups. These two parameters are determined as the right foot hits the surface and is not expected to change because it is something that is planned before stepping onto the different surfaces. There was a significant **lower** value for the single support durations for surface with dowels and the surface with ridges as compared to the control surface with the average difference of 30 ms ($p<0.0001$). For the double support durations, only the compliant sponge surface showed significantly **lower** value as compared to the control surface with the average difference of 55 ms ($p<0.0001$). The stride velocity of the steps on the experimental surfaces showed significantly **higher** values as compared to the control surface with the average difference of 10 cm/s ($p<0.0001$).

***Table 2a** – Mean values for the spatiotemporal gait parameters for the third, fourth and the fifth step on surfaces 1 to 6.*

3^{rd} **Step** (Right)		4^{th} **Step** (Left)	5^{th} **Step** (Right)
Parameter	Surface Means (S.E)	Surface Means (S.E)	Surface Means (S.E)
Stance Duration (sec)	1 - 0.662 (.008) 2 - 0.638 (.008) 3 - 0.643 (.008) 4 - 0.648(.008) 5 - 0.610(.008) 6 - 0.614 (.008)	1 - 0.641(.012) 2 - 0.604(.012) 3 - 0.621(.012) 4 - 0.613(.012) 5 - 0.611(.012) 6 - 0.605(.012)	N.A.
Swing duration (sec)	1 - 0. 453 (.005) 2 - 0. 416 (.005) 3 - 0. 424 (.005) 4 - 0. 419 (.005) 5 - 0. 468 (.005) 6 - 0. 421 (.005)	1 - 0. 458(.006) 2 - 0. 439(.006) 3 - 0. 432(.006) 4 - 0. 427(.006) 5 - 0. 469(.006) 6 - 0. 430(.006)	1 - 0.466(.007) 2 - 0.427(.007) 3 - 0.436(.007) 4 - 0.429(.007) 5 - 0.443(.007) 6 - 0.450(.007)
Single Support Duration (sec)	1 - 0. 458 (.006) 2 - 0. 439 (.006) 3 - 0. 432 (.006) 4 - 0. 427 (.006) 5 - 0. 469 (.006) 6 - 0. 430 (.006)	1 - 0. 467(.007) 2 - 0. 427(.007) 3 - 0. 436(.007) 4 - 0. 429(.007) 5 - 0. 444(.007) 6 - 0. 451(.007)	N.A.
Double Support Duration (sec)	1 - 0. 205 (.007) 2 - 0. 200 (.007) 3 - 0. 211 (.007) 4 - 0. 222 (.007) 5 - 0. 149 (.007) 6 - 0. 184 (.007)	1 - 0. 179(.009) 2 - 0. 183(.009) 3 - 0. 192(.009) 4 - 0. 190(.009) 5 - 0. 162(.009) 6 - 0. 158(.009)	N.A.
Step Length (cm)	1 - 74.930 (.941) 2 - 75.150 (.941) 3 - 75.369 (.941) 4 - 75.334 (.941) 5 - 76.939 (.941) 6 - 72.982 (.941)	1 - 70.294(.972) 2 - 75.454(.972) 3 - 75.547(.972) 4 - 73.553(.972) 5 - 72.327(.972) 6 - 72.296(.972)	1 - 73.413(.988) 2 - 74.645(.988) 3 - 75.813(.988) 4 - 73.063(.988) 5 - 76.991(.988) 6 - 71.796(.988)
Step Width (mm)	1 - 75.830(.944) 2 - 75.900(.944) 3 - 76.727(.944) 4 - 76.114(.944) 5 - 77.599(.944) 6 - 73.390(.944)	1 - 72.266(.974) 2 - 76.914(.974) 3 - 76.762(.974) 4 - 74.407(.974) 5 - 73.172(.974) 6 - 73.631(.974)	1 - 74.424(.909) 2 - 75.855(.909) 3 - 76.258(.909) 4 - 74.436(.909) 5 - 78.401(.909) 6 - 73.047(.909)

Table 2b *– Results of the univariate statistical analysis of the main effects of surface on spatiotemporal Gait parameters for the third, fourth and the fifth step. Level of Significance (F value)*

Parameters	*1st Step on Surface(Right) 3rd*	*Next Left Step 4th*	***Subsequent Right step 5th***
Stance Duration	**0.0001*** (6.175)	**0.6** (0.784)	NA
Swing Duration	**0.0001*** (18.155)	**0.001*** (8.076)	**0.001*** (4.384)
Step Duration	**0.0001*** (15.749)	**0.002*** 4.037	**0.001*** (4.733)
Single Support Duration	**0.0001*** (8.076)	**0.001*** (4.610)	NA
Double Support Duration	**0.0001*** (13.440)	**0.04*** (2.454)	NA
Stride Velocity	**0.0001*** (6.892)	**0.0001*** (9.127)	**0.0001*** (7.923)
Step Length	**0.1** (1.813)	**0.001*** (4.385)	**0.004*** (3.735)
Step Width	**0.06** (2.226)	**0.003*** (3.891)	**0.002*** (4.201)

(* Significant difference)

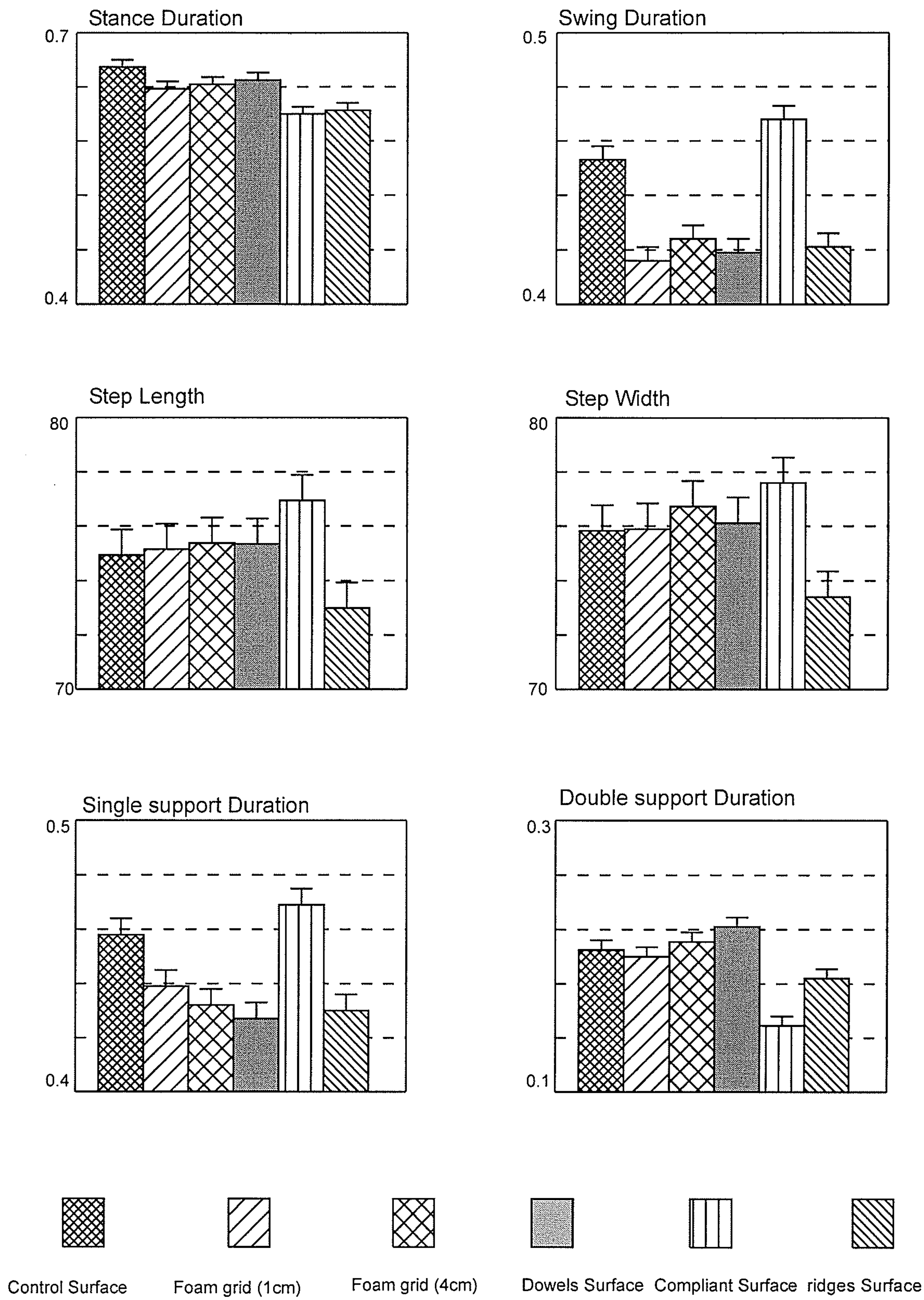

Figure 2: Group means and Standard error of means for the Spatial-temporal Gait parameters for the first step on the surface. For the first and last row of the histograms Y axis is the time scale in seconds. For the middle row the Y axis represents the distance in cm.

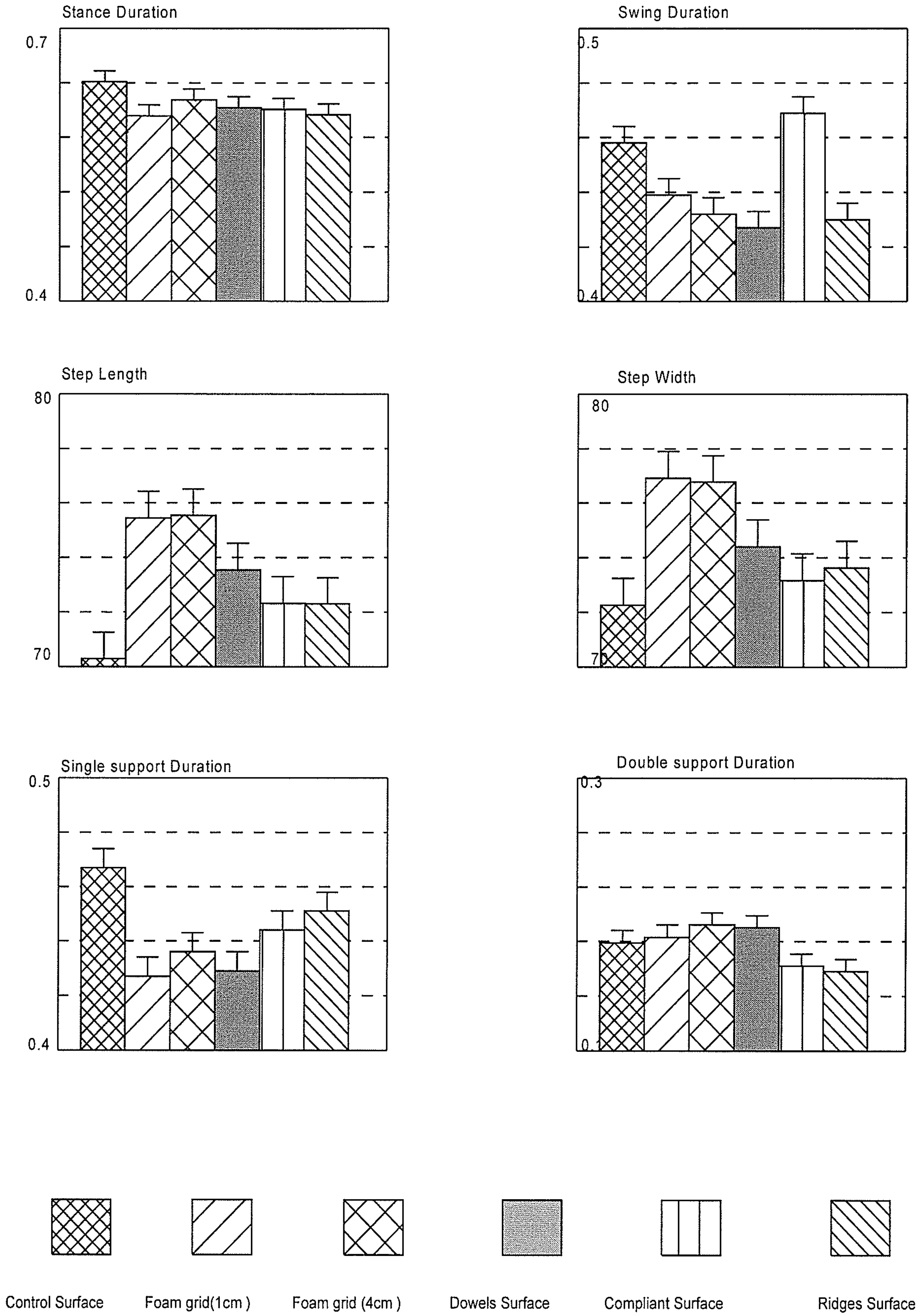

Figure 3: *presents the group means and standard error of means for the spatial-temporal gait parameters for the 4th step (left step after the surface)*

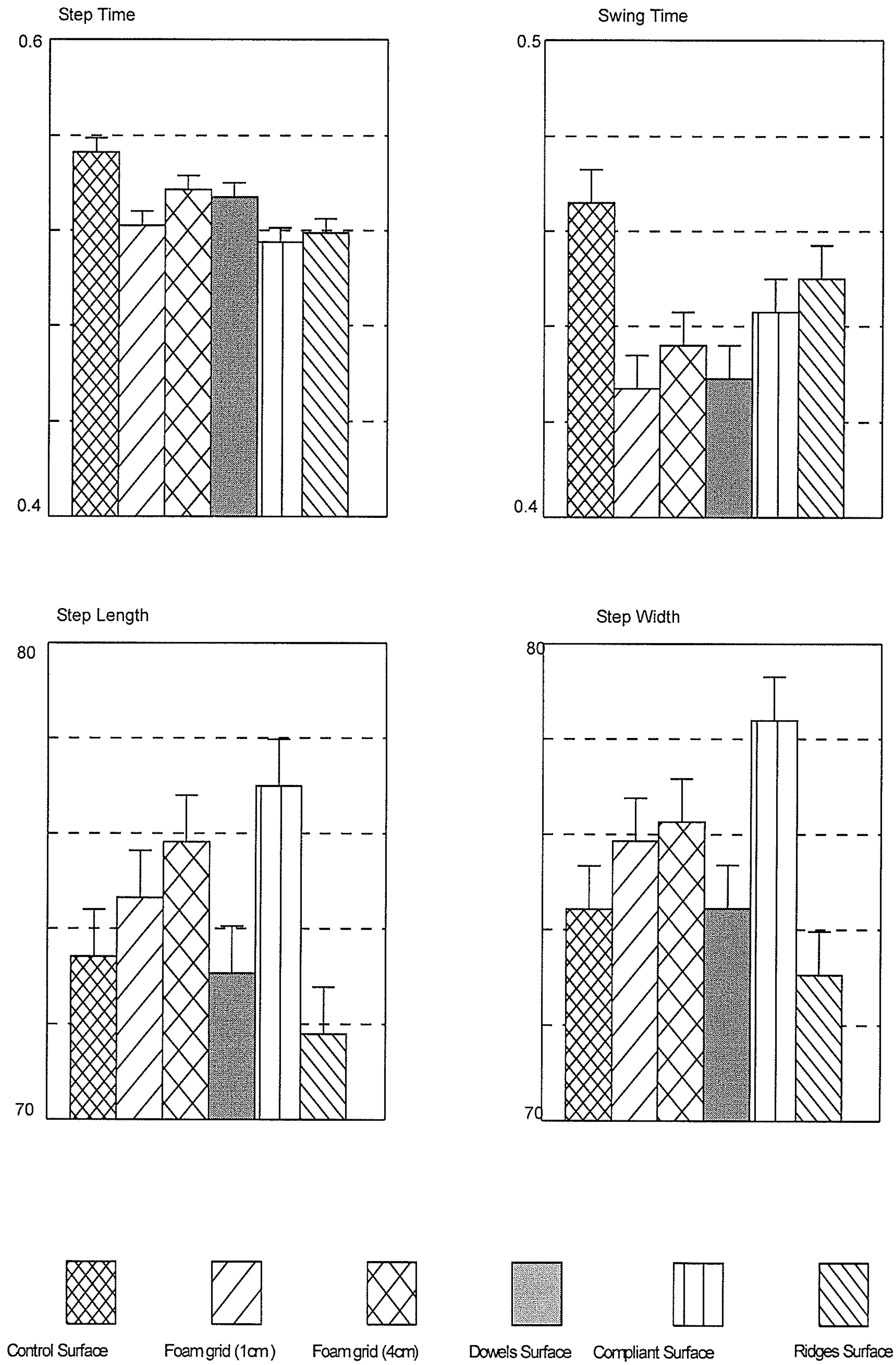

Figure 4 *presents the group means and standard error of means for the spatial-temporal gait parameters for the 5^{th} step (Second right step after the surface).*

Figure 3 presents the group means and standard error of means for the spatio-temporal gait parameters for the 4th step, *which was the first left step after the surface.*

The *stance duration* did not show any significant difference for the different surface conditions. For the *swing durations* the surface with compliant grid (4cm gap), the surface with dowels and the surface with ridges showed highly significant **lower** values as compared to the control surface with the average difference of 28 ms ($p < 0.001$). For *single support durations,* surfaces with foam grids (1cm gap) and the surface with dowels showed significant **lower** values from the control surface with the average difference of 36 ms ($p < 0.001$). For the *double support duration*, the compliant surface and the surface with ridges had significantly **lower** values from the control surface with the average difference of 20 ms ($p<0.04$). *Step length* and *step width* was significantly **higher** for both the foam surfaces with grid as compared to the control surface with the average difference of 5cm and 4.57 cm respectively ($p<0.001$ & $p< 0.003$).For this step the value for *stride velocity* was significantly **higher** for all the experimental surfaces as compared to the control surface with the average difference of 11.3 cm/s ($p<0.0001$).

Figure 4 presents the group means and standard error of means for the spatial-temporal gait parameters for the 5th step *(2nd right step after the surface)*. For the variable *step duration*, the surface with foam grid (1cm gap), the exercise mat, and the surface with ridges showed significantly **lower** values from the control surface with the average difference of 34 ms ($p<0.001$). For the *swing duration*, foam grid (1cm gap and with 4cm gap), and the surface with dowels were significantly **lower** as compared to the control surface with the average difference of 35 ms ($p<0.001$). For both *step length* and *width* significantly **greater** values were found for the exercise mat as compared to the control surface with the average difference of 3.6 cm and 4 cm respectively ($p<0.004$ & $p<0.002$). For this step the *stride velocity* was significantly **higher** for all the experimental surfaces as compared to the control surface with the average difference of 10cm/s ($p<0.0001$).

6.2.2 Plantar pressure parameters

Figure 5 presents the raw data for the pressure patterns for the stance phase of the first right step on the test surface for a subject who was unstable (A) and a subject who was stable (B). Effect of three surface conditions has been presented: the control surface, surface with dowels and the surface with ridges. The peak force occurred in the heel region in the early part of stance phase and in the forefoot region it occurred at a later time. This is a pattern that is similar to a normal heel – toe gait pattern.

There is little difference in the pressure patterns in the heel region for subjects compared to each other. For the medial forefoot region the control surface shows higher pressure than the two experimental surfaces in both the subjects. For the lateral forefoot region the surface with ridges shows the highest pressure in both the subjects. The last two graph panels present the COP traces in the medio-lateral and anterior-posterior directions. As evident, the unstable subject A had higher deviation of the COP in the medio-lateral direction as compared to subject B.

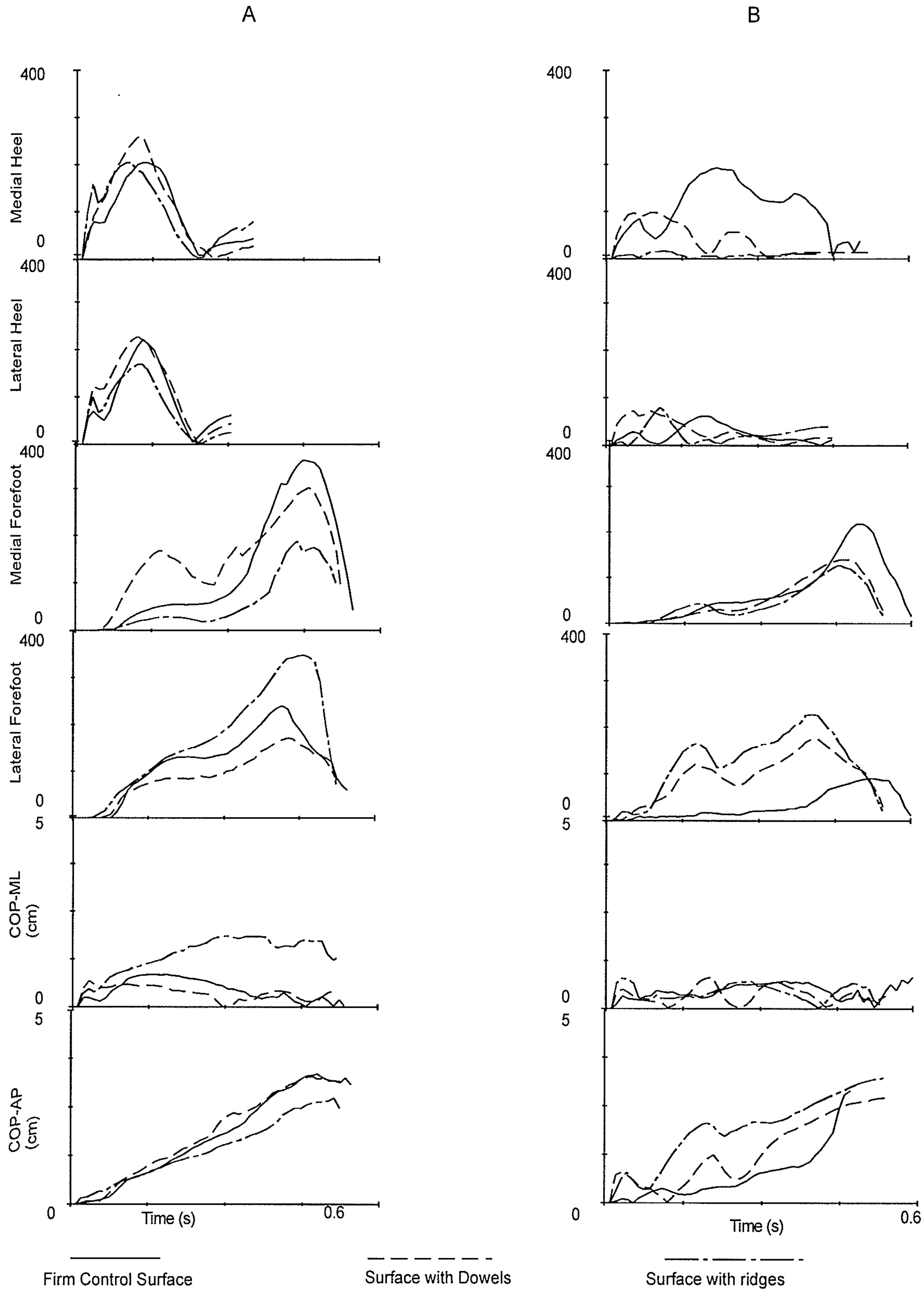

Figure 5 – Plots of Plantar pressures and COP for first step on the surface (right stance) for Subject A (unstable) and B (stable). First four rows represent the total plantar pressures (units: psi) and the last two are the COP traces (units cm). All curves have been offset to zero for display purposes. Effect of three surfaces has been illustrated – Firm control surface, surface with dowels and the irregular surface with ridges

6.2.2a – Pressure time integrals: These values were only measured for the right foot. **Figure 6** presents the group means and standard errors of mean for the pressure - time integral for the first step on the surface for the six different surfaces. The pressure time integral for the **medial heel** were significantly **lower** for all the experimental surfaces as compared to the control surface with the average difference of 1727.3 units ($p<0.05$). For the **lateral heel region** surface with dowels and the surface with ridges had significantly **lower** values than the control surface with the average difference of 2002.2 units ($p\leq0.001$) whereas for the **medial forefoot region** this value was significantly **higher** for the experimental group as compared to the control surface with the average difference of 1205.7 units ($p\leq0.5$). For the *next right step* the lateral heel was the only area which showed the significantly lower values from the control surface. In general, the **pressure time integral** for the heel region had **lower** values for experimental surfaces and the forefoot had higher values as compared with the control surface.

Table 3 shows the level of significance and the F values for the pressure time integral in the different areas of foot: the medial heel, the lateral heel, mid-foot, medial forefoot and the lateral forefoot.

6.2.2b – Peak pressure: **Figure 7** presents the group means and standard error of mean for the **peak pressure** for the different zones of the foot which showed a significant difference. For the *first step on surface (right)*, a significant difference was observed only for the medial forefoot region. For this area the peak pressure for the surface with ridges was significantly **lower** than the control surface with the average difference of 3.2 psi ($p\leq 0.05$). For the *next right step on the surface*, the peak pressure for both the **medial heel** and the **lateral heel** showed significant **lower** values for the surface with ridges as compared to the control surface with the average difference of 3.7 psi and 4.6 psi ($p \leq 0.01$) respectively. In general the **peak pressures** values were **lower** for experimental surfaces as compared to the control surface. **Table 4** shows the level of significance and the F values of peak pressure in different areas of the foot.

6.2.2c – Maximum excursion of COP: **Table 5** presents the maximum excursion of center of foot pressure in ML and AP plane. There was no significant difference observed in this parameter for the first and second right step on the surface.

6.2.3 Maximum excursion of Centre of Mass in Medio-lateral direction

Figure 8 presents the raw plots of Center of mass during the first step on the surface (right) for Subject A (unstable) and B (stable) in M-L and A-P direction. Effect of three surfaces has been illustrated – control surface, surface with dowels and the surface with ridges. The COM in the medio-lateral direction starts to rise as the subject takes a step and reaches a peak in the middle of the step and then falls back to the same level at the end of the step. This reflects the normal sine curve followed by the centre of mass in the medial-lateral direction during walking. The COM displacement in the anterior posterior direction keeps on increasing as the subject progresses. **Table 6** shows the result of univariate analysis for maximum displacement of COM in ML direction. There was no significant difference observed in this value between control surface and the experimental surfaces.

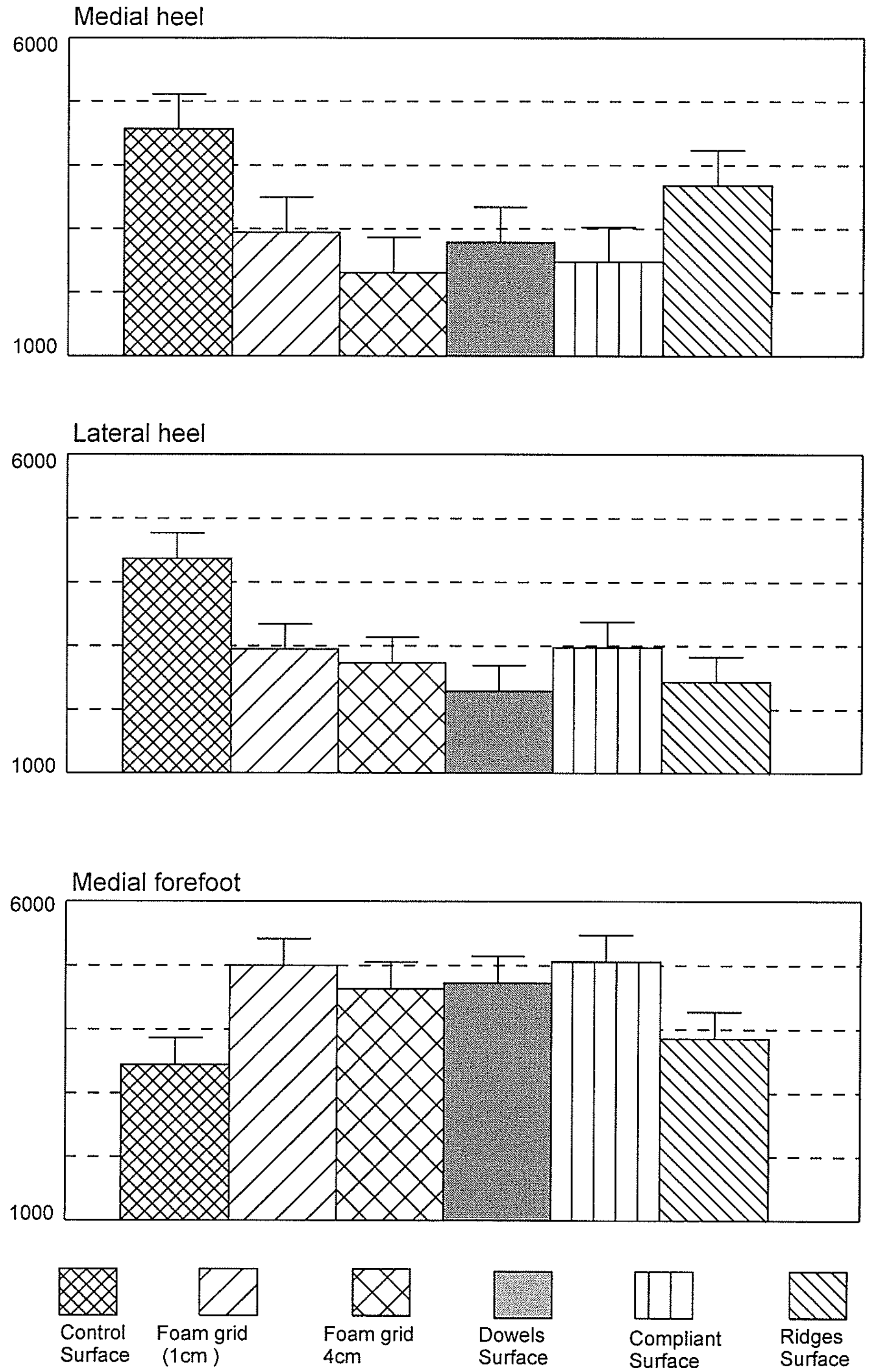

Figure 6: Group means and standard error for the Pressure - time Integral for the first step on the surface for the six surfaces. These three zones of the foot showed the significant difference. Y axis represents pressure time integrals in units *psi* times *time.* *(Psi = pounds per square inch)*

***Table 3** – Result of the univariate statistical analysis for the Pressure time integral for different zones of foot: Level of Significance (F Value).* (* Significant difference)

AREA	*Medial Heel*	*Lateral heel*	*Mid-Foot*	*Medial Forefoot*	*Lateral Forefoot*
1st Step on Surface(Right)	**0.05*** (2.381)	**0.008*** (3.478)	**0.5** (0.860)	**0.04*** (2.453)	**0.06** (2.253)
Next Right Step	**0.5** (.832)	**0.04*** (2.570)	**0 .6** (0.681)	**0. 7** (0.621)	**0.4** (1.041)

***Table 4** – Result of the univariate statistical analysis for Peak pressure for different zones of foot: Level of Significance (F Value)*

AREA	*Medial Heel*	*Lateral heel*	*Mid-Foot*	*Medial Forefoot*	*Lateral Forefoot*
1st Step on Surface(Right)	**0.1** (1.763)	**0.3** (1.285)	**0.4** (1.073)	**0.05*** (2.424)	**0.8** (0.431)
Next Right Step	**0.02*** (3.111)	**0.0001*** (6.469)	**0.3** (1.250)	**0.4** (1.024)	**0.4** (0.966)

***Table 5** – Result of the univariate statistical analysis for maximum excursion of COP in M-L and A-P direction: Level of Significance (F Value)*

	COP X *(Medio-lateral)*	COP Y *(Antero-posterior)*
1st Step on Surface(Right)	**0 .6** (0.732)	**0.053** (2.317)
Next Right Step	**0.8** (0.459)	**0.8** (0.514)

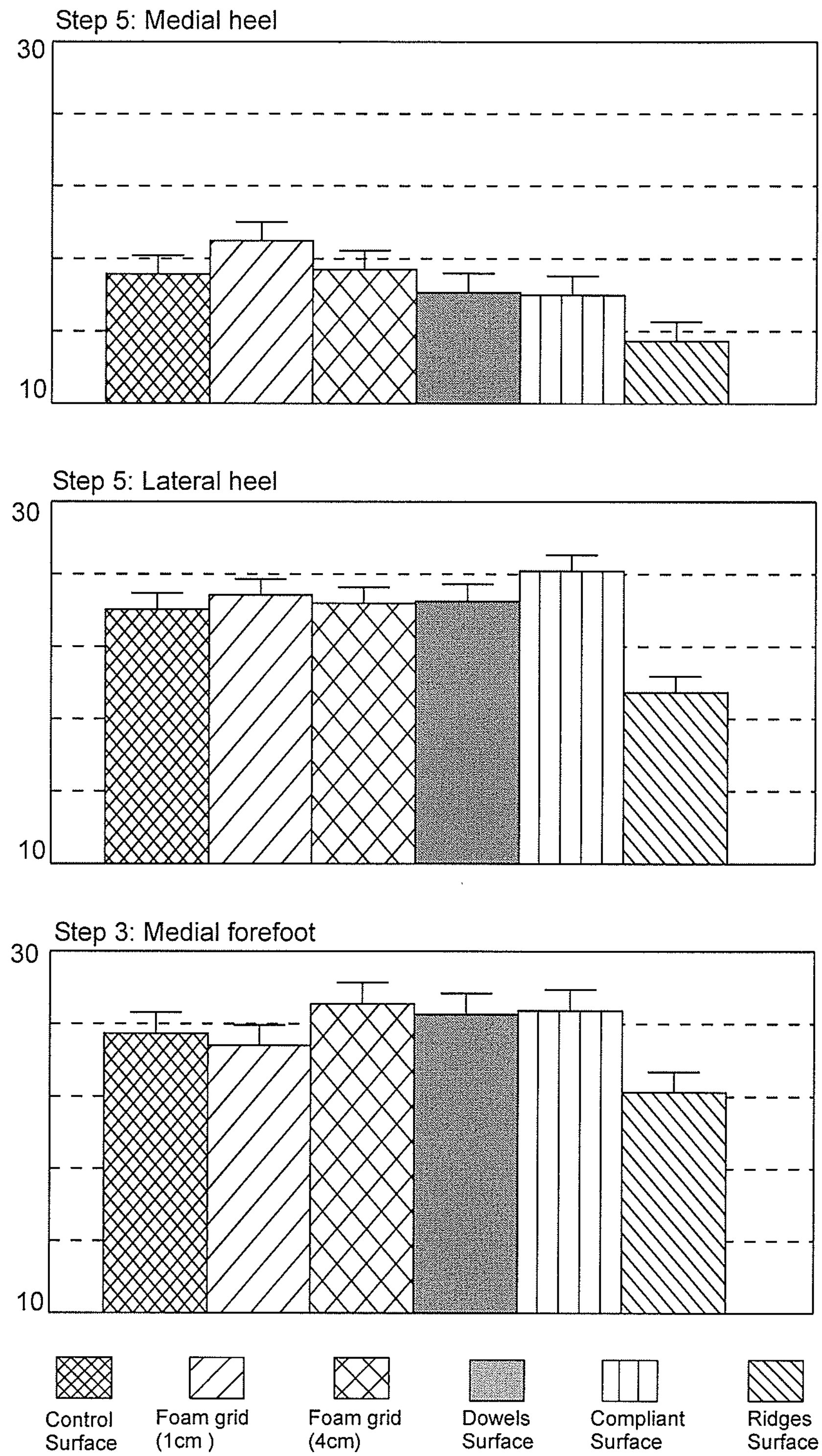

Figure 7 Histograms depicting the group means and standard error for the Peak pressure for the six surfaces for different zones of the foot which showed the significant difference. Unit for Y axis is psi (pounds per square inches)

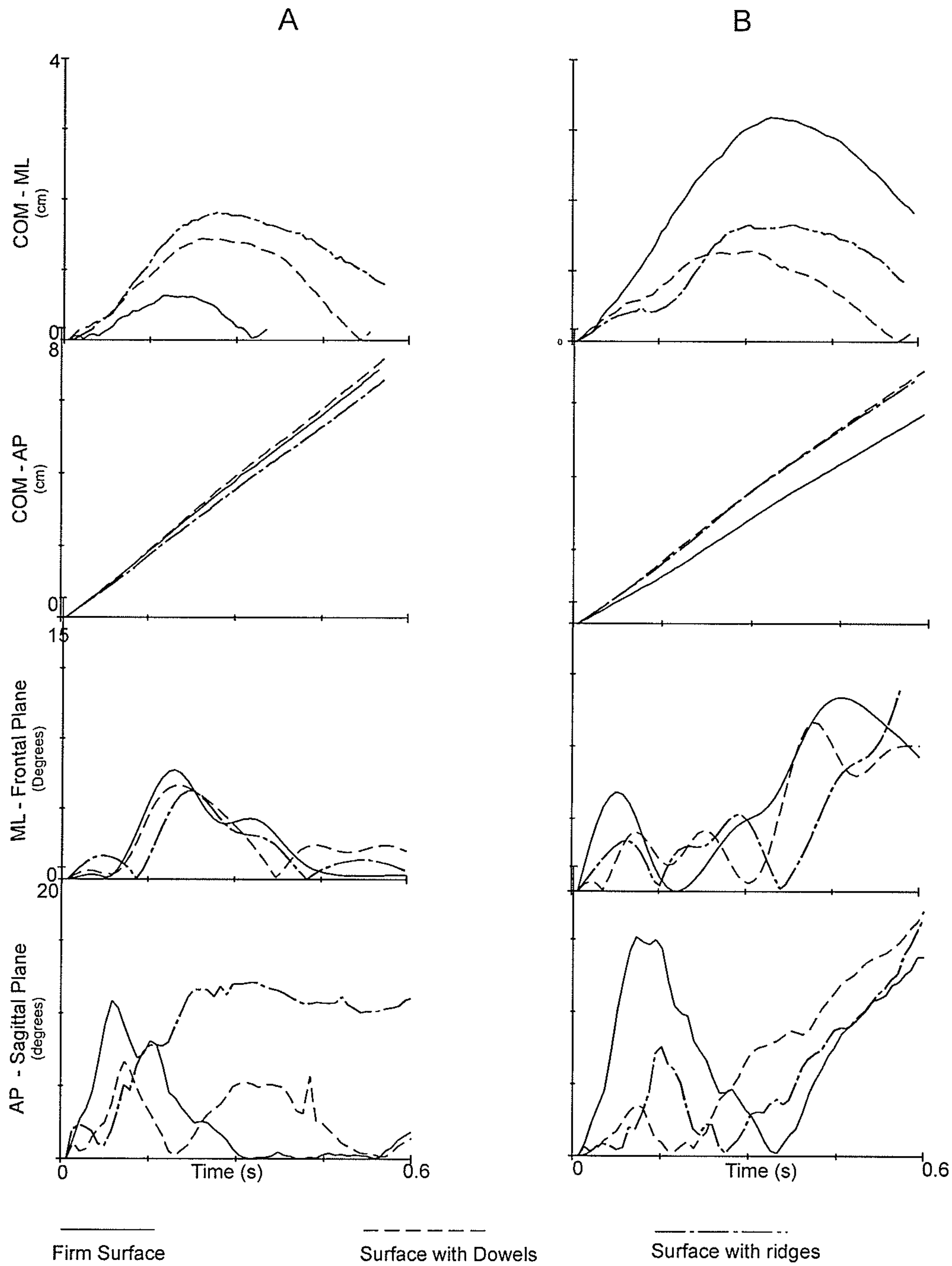

Figure 8 Plots of Center of mass and ankle angle displacements during the first step on the surface (right) for Subject A (unstable) and B (stable) in M-L and A-P direction. All curves have been offset to zero for display purposes. Effect of three surfaces has been illustrated Firm, with dowels and the irregular surface with ridges.

6.2.4 – Correlation analysis of COM and COP:

There were no significant statistical differences in peak cross- correlation r-values of COM and Cop trajectories among the different support surfaces. The r-values ranged from 0.90 to 0.99.

6.2.5 – Correlation analysis of ankle angle and COM:

Figure 8 presents the raw plots of ankle angle displacements in ML and AP plane. The patterns of the waveform for the ankle angle are similar for the experimental and the control surface except for subject A in the AP plane where the surface with ridges shows significant amount of deviation from the control surface. This difference was observed for this subject specifically.

Figure 9 presents the group means and standard error for the cross-correlation r-values values for the six surfaces. As evident from the figure the control surface has the highest peak correlation coefficient value ($r = 0.84$) and the least for the dowel surface ($r = 0.38$). **Table 7** presents the level of significance and the F value from the result of the univariate statistical analysis for the for the peak r-values of the first step on the surface. There was a highly significant difference observed between the different surface conditions for the peak r-value for the ankle angle and COM in the medio-lateral direction ($P< 0.0001$). There was no significant difference in the anterior-posterior direction. The cross correlation values did not show any significant difference in either of the plane.

Table 6 *– Result of the univariate statistical analysis for maximum excursion of COM in ML direction: 1st Step on the Surface Level of Significance (F Value)*

	COM X *(Medio-lateral)*
1st Step on surface(Right)	**0.6** (0.702)

Table 7 *– Result of the univariate statistical analysis for peak correlation coefficient (r value) and cross correlation between Ankle angle & Centre of Mass for 1st Step on the Surface: Level of Significance (F Value)*

	Antero-posterior	*Medio-Lateral*
Correlation Coefficient	**0.2** (1.439)	**0.0001*** (5.768)
Cross Correlation	**0.6** (0.774)	**0.7** (0.610)

Table 8 *– Result of the univariate statistical analysis for maximum correlation coefficient (r value) between Trunk & Leg Accelerations Signals for the left swing leg after the surface: Level of Significance (F Value).*

	Antero-posterior	*Medio-Lateral*
Correlation Coefficient	**0.6** (0.738)	**0.2** (1.614)
Cross Correlation	**0.4** (0.993)	**0.6** (0.802)

(* Significant difference)

6.2.6 – Correlation analysis between trunk and leg acceleration signals:

Table 8 shows the group averages of the peak correlation coefficient and the cross correlation values for the trunk and shank acceleration signals for the first left step after the surface. There was no significant difference observed for the different surface conditions for these variables.

Figure 10 presents the raw acceleration signals of trunk and shank for the first left step after stepping on the surface. For subject A the trunk acceleration signal on ridges surface shows larger deviations from the control surface whereas for the subject B the deviations are significantly less. Ankle acceleration signals for subject A shows larger amount of deviation for the experimental surfaces when compared to that for the subject B.

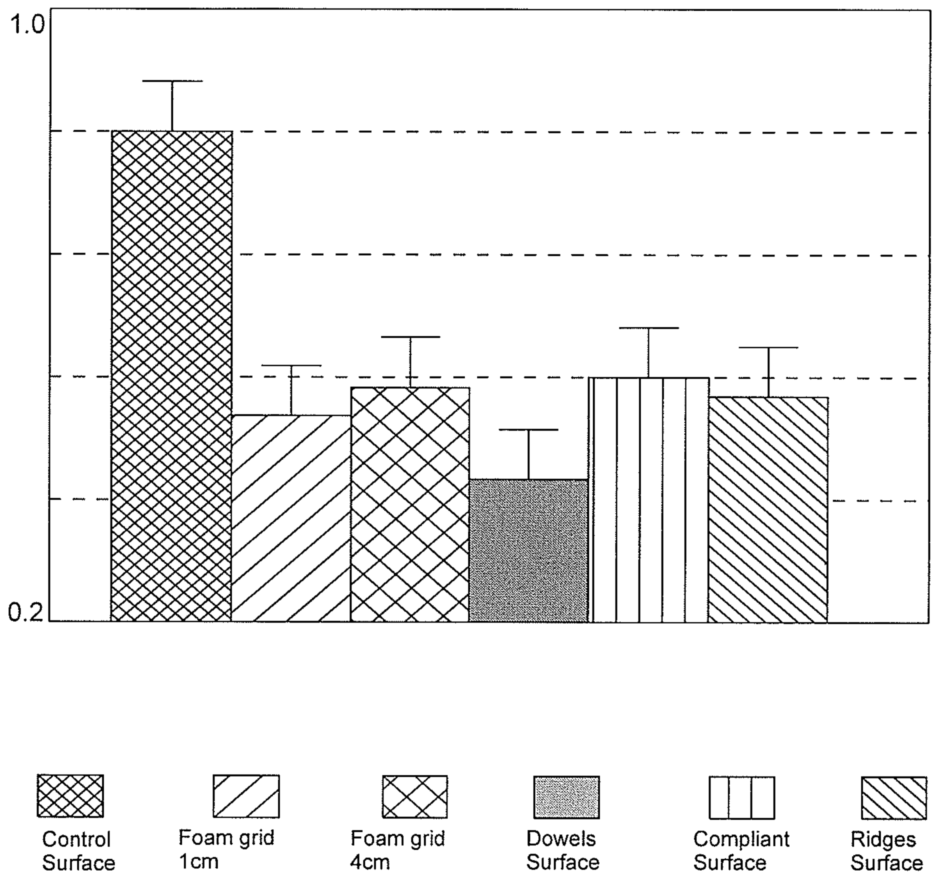

Figure 9 Group means and standard error for the peak correlation coefficient values for the six surfaces for the Center of Mass and Ankle angle in the Medio-lateral direction. Y axis represents the values for the correlation coefficient.

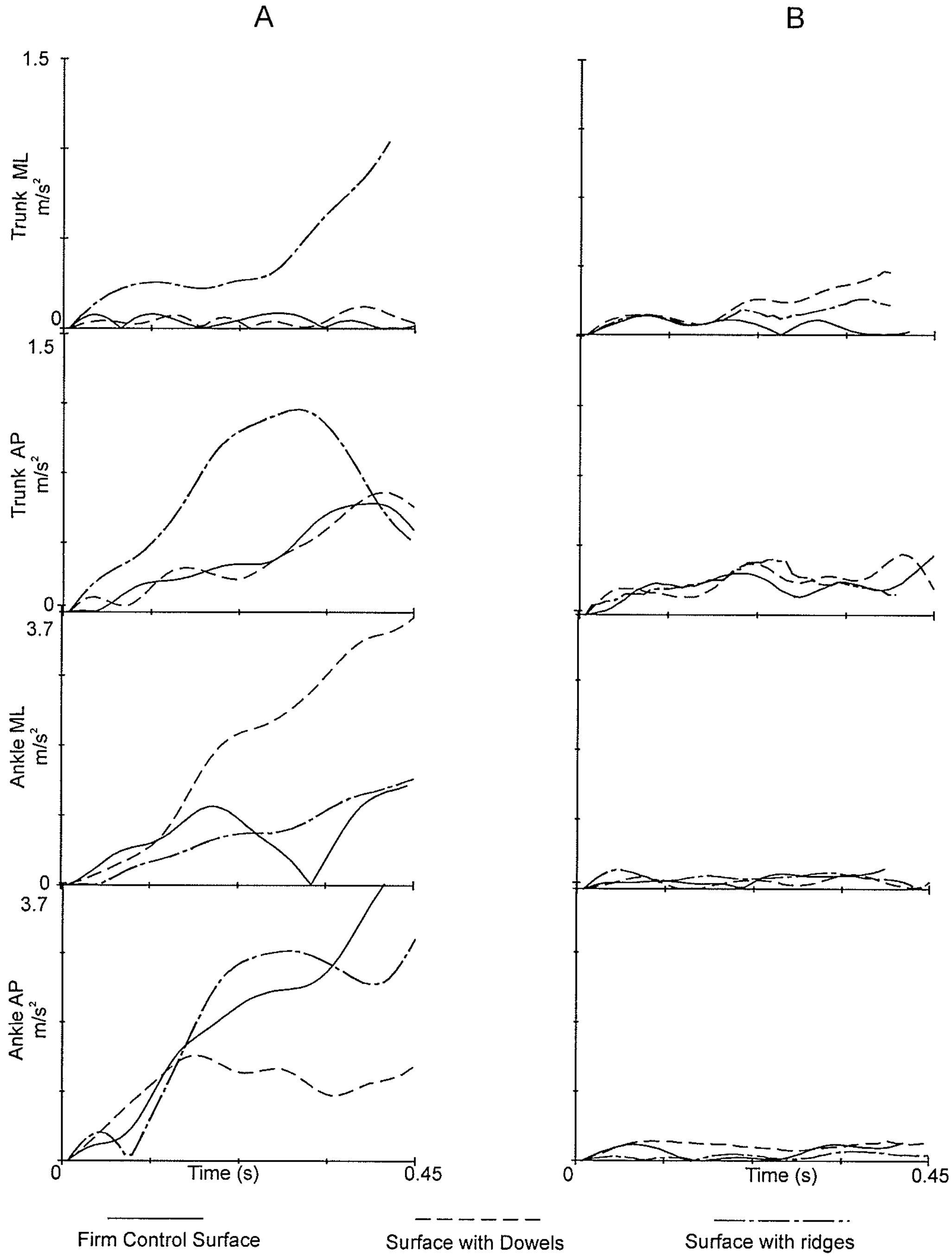

Figure 10 Plots of Trunk and Shank acceleration signals for the step after the surface (Left swing leg) for Subject A (unstable) and B (stable) in M-L and A-P plane. All curves have been offset to zero for display purposes. Plots of three surfaces have been presented - Firm control surface, surface with dowels and the surface with ridges.

7. DISCUSSION

The main purpose of the study was to evaluate balance control while walking on firm flat surfaces as compared to irregular and compliant surfaces.

The main findings of the study:

- There were no changes observed in any spatial-temporal gait parameters prior to stepping on the surfaces.
- Most of the spatiotemporal variables, such as the stance time, swing time, and step length; step width and stride velocity; as well as single and double support durations did significantly change while walking on the experimental surfaces.
- The heel region of the foot had significantly lower pressure time integral values whereas the forefoot had significantly higher pressure time integral values while walking on the experimental surfaces as compared to the control surface.
- Peak plantar pressures were significantly lower in the heel and the forefoot region while walking on the different experimental surfaces as compared to the control surface.
- The peak value of correlation coefficient carried out between ankle angle and center of mass was significantly higher for the control surface as compared to the different experimental surfaces.
- There was no significant difference in the whole body centre of mass between the control surface and the experimental surfaces.
- The whole body coordination was not affected while walking on the irregular and/or compliant surfaces.

7.1 Altered surfaces cause significant changes in the spatiotemporal gait parameters.

In this study, a 5 m walkway was used in which the middle 1-3 m was either a control surface or an irregular / compliant surface, not known to the participants. We did not observe any signs of adjustments in step velocity, timing or distance gait parameters in anticipation of stepping on any of the experimental surfaces.

Studies have been carried out on young and older adults that showed that walking on irregular surfaces resulted in an increase in the stride length, step time variability and the cadence of young, healthy subjects. They did not find any difference in the gait speed between the regular and irregular surfaces for the younger subjects. In other studies, the older subjects had slower walking speed, increased step time and width, shorter step length, and greater step timing variability for both the surfaces as compared to their younger counterparts (Menz et al 2003a, b; Theis et al 2005). In these studies, subjects took separate walking trials on a level corridor forming the regular surface and on a continuous walking surface with irregular shaped objects placed randomly throughout. After stepping on the irregular surface, subjects knew that the entire platform would be irregular. This knowledge of the surface condition could have led to a preplanned walking pattern and subjects could have adopted a cautious walking pattern on a known irregular and potentially disturbing surface. In these studies, the gait parameters were averaged over the entire trial. Averaging the parameters obscures the balance reaction which occurs when subjects first encounter the irregular surface. In our study, we had randomly presented six different surfaces to the subjects and then analyzed the first step on the surfaces and the subsequent two steps. Analyzing the steps separately provides a better idea of the balance reactions to an unexpected exposure to an irregular or compliant surface. Moritz and Farley (2004) showed that anticipation of an altered surface causes changes in the joint angle even before the beginning of the altered surface. The change was seen in terms of increase in the knee joint flexion. This was not found when subjects landed on an unexpected altered surface. This study shows that subjects become more cautious when they are aware of a change or a disturbance in the surface condition. Our findings demonstrate that the temporal parameters such as the stance time, the swing time, the single and the double support durations had significantly **lower** values on the experimental surfaces as opposed to the control surface, whereas the spatial parameters such as the step length, the step width and the stride velocity had significantly **higher** values on the experimental surfaces. This corresponds with the fact that after encountering the experimental surfaces, subjects increased their step length and speed to cross them with an increased base of support to restore stability. Walking rhythm was affected by the mechanical disturbance which caused a deviation or a stumble (small to medium,) requiring a timely compensatory reaction to prevent a large deviation such as a stop in progression or a fall. Maclellan and Patla (2006) found that walking on a large foam mat increased the step length and width of the subjects but did not affect the medio-lateral COM. This finding is consistent with our study.

The following studies have shown that alteration at the level of the foot plays an important role in gait patterns. However in these studies, the factor of uncertainty was not incorporated as the subjects had the same walking condition throughout the walking trial. They focused on average response during the walking trial as opposed to our study where the first exposure to the changed surface condition and the subsequent two steps were analyzed separately. Nurse et al (2005) studied the role of afferent cutaneous feedback from the plantar aspect of foot on gait performance measures. Subjects walked with either a smooth or a textured shoe insert. The textured shoe insert was used in order to alter the sensory input of the right foot and to evaluate whether this change at the level of the foot can modify gait rhythm and pattern. The foot was found to be significantly more plantar flexed at heel strike in the textured shoe insert group. There was a significant reduction in the soleous and tibialis anterior activity in the group with the textured shoe insert as opposed to the group with the smooth shoe insert. Eils et al (2004) studied the effect of reduced plantar sensation on the gait pattern. They found significant changes in the walking pattern in the group with reduced plantar sensation in terms of joint kinematics at ankle, knee and hip and a reduction in the vertical ground reaction force. This study shows how altered sensation from the plantar aspect of the foot modifies gait parameters. McDonnel et al (2000) induced partial foot anaesthesia with the help of local anaesthetics and studied the effect on gait speed while subjects walked on a compliant foam walkway. They found significant reduction in mean gait velocity for subjects having anaesthesia as compared to the control group. This study shows that the sensory feedback from the mechanoreceptors in the feet is an important determinant of motor output during walking. This is consistent with the findings of our study where altered support surfaces caused significant changes in the gait parameters.

7.2 Influence of altered surface on plantar pressure parameters

The pressure time integral (PTI) and the peak pressures (PP) are considered important parameters for evaluating the distribution and changes in plantar pressure. These variables form an important clinical consideration for the analysis of therapeutic footwear (Erdemir et al. 2005, Chen et al. 2003, Xu H et al 1999). In our study, the PTI for the experimental surfaces had significantly **lower** values for the **heel** region as compared to the control surface. The **medial forefoot** region showed a significant **increase** in the PTI values for the experimental surface. The peak plantar pressures for the **forefoot** and the **heel** region were significantly **lower** for the experimental surfaces as compared to the control surface. The mid-foot region did not show any significant difference.

Lower values of the PTI at the heel reflects that after stepping on the experimental surfaces, subjects raised their heel immediately and their foot progressed quickly to mid-stance which resulted in spending less time in the heel region. Decrease in the peak pressure could be attributed to the compliant nature of three of the foam surfaces and the reduced surface area for contact of the dowels and ridged surface. Studies support that hard surface tends to increase the plantar pressure as compared to the compliant surfaces. Mohamed et al. (2005) and Burnfield et al (2004) carried out studies to compare the plantar pressure parameters on different terrains like grass, carpet and concrete surfaces. Walking barefoot was compared to walking with shoes on. These studies concluded that PP, PTI and maximum mean pressure were significantly higher when walking barefoot on the hard concrete surface as compared to the compliant grass or carpet. Similar to our study, these changes were significant in all the areas of the foot except for the mid foot region.

Reducing plantar sensation has a significant affect on walking ability and this is also shown by studies which evaluate walking performance in patients with substantial peripheral Neuropathy due to diabetes (Nurse and Nigg, 2001; Eils et al., 2002).

In our study, the reduced pressure parameters in the heel and decreased temporal gait values indicate that the sensory information has been reduced or distorted. A similar effect has been studied while standing on the foam sponge and was found that the foam base modifies the ground reaction forces in an unpredictable manner (Allum et al 2002). The surface being compliant can not accept the normal forces from the feet as the body sways and moves. The compliant surface thus introduces uncertainty into the system that will result in distortion and delay in interpretation of the signals coming from cutaneous sensors of the feet. The mechanical effect of the irregular or the compliant surfaces can be detected by the mechanoreceptors of the foot only after the foot contacts the support surface (Maclellan and Patla 2006). This will in turn make it difficult for participants to estimate and determine necessary adjustments required to control the motion and position of COM relative to the support surface.

A higher value of PTI at the **forefoot** region suggests that subjects had to apply significantly higher pressure for pushing off from the irregular and compliant surfaces. Our result is different from the study done by Eils et al (2002) where they did not find any significant difference in the PTI. This difference could be due to the fact that in the study by Eils et al. (2002), a reduction in plantar sensation was induced by application of ice whereas in our study, the base of support was compliant or irregular. This change in support surface could have altered the cutaneous input from the foot and would have made subjects more cautious to make a steady push off for good stability.

7.3 Dynamic stability during a steady state walk on irregular surfaces.

Stability while walking is dependent on controlling whole body COM within a moving base of support. For the present study, the index of stability was defined as the maximum correlation value between COM and COP. However, the results of this correlation value were inconclusive because there was virtually no difference in the trajectory of the COP positions among the different surfaces.

The cross correlation analysis between **ankle angle** and **COM** was also performed to assess the dynamic stability during walking on the irregular and compliant surfaces. The peak value of correlation coefficient was computed for these two variables in ML as well as AP plane. The experimental surface had significantly lower cross correlation values indicating that the subjects were less stable as compared to the control surface.

The control surface had the highest 'r' value of (0.84) and the dowel surface had the least 'r' value (0.38) which indicated least stability on that surface in the medio-lateral direction. Gatev et al. (1999) carried out the cross correlation analysis of the centre of mass and centre of foot pressure on different surface conditions in stance. They found that with the narrower support surface, the peak correlation value decreased showing that subjects become less stable with a narrower base of support. In our study, we did not alter the width of the base of support but made it irregular or compliant. Results indicate that the experimental surfaces did introduce instability in the walking task as opposed to the control surface.

7.4 Coordination between body segments while walking on irregular surfaces

Swing leg trajectory and consequently the foot placement is a primary means of controlling stability (Patla 2003). Cross correlation was carried out on the acceleration signals for the left shank segment and the trunk segment for the step after the surface. This measure was taken as an index of motor coordination and was computed for the left swing phase, the first left step after contact with the different surfaces. The results did not show any significant difference in this index of motor coordination for the different surface conditions. The experimental surfaces did produce a stumble, but the motion of the different body segments continued to be highly correlated and synchronized, which indicates continued coordination during the compensatory balance reactions.

This leads to the inference that the disturbance due to the experimental surfaces was not significant enough to cause a change in the coordination of the body segments. Even though the index of stability shows that the subjects were less stable on the experimental surfaces, this parameter reflects that they did not become uncoordinated after stepping on any of the experimental surfaces.

Cross correlation studies have analyzed the body segments during stance on different support surfaces. Mesure et al (1997) evaluated postural performance by cross correlating the head and hip acceleration signals. Subjects were standing either on firm or compliant support surfaces. They found that the postural performance on the foam rubber surface was very significantly impaired with respect to firm support surface. The foam surface induced uncertainty in the system and thus coordination was affected. Nardone et al (2000) carried out cross correlation between head and hip segment for young and older subjects standing on the sinusoidally moving platform. Increase in the frequency of the platform resulted in less coupling of the head and hip segment. In both the above studies, perturbation of support surfaces altered the coordination of body segments while in stance. In our study, there was no significant change in the coordination of body segments. This might be due to the fact that the subjects had a momentum of walking. The central nervous system had the ability to deal with the disturbance produced by the experimental surfaces and thus not altering the coordination between body segments. Stepping on to the compliant and/or irregular surface, did challenge stability but was reflected more in the spatiotemporal gait parameters such as an increase in the step width and step length and the whole body co-ordination was not compromised.

8. CONCLUSION

The purpose of the present study was to evaluate the balance requirements of steady-state walking on compliant and irregular surfaces that would be typical of outdoor terrains. The foot is the contact point during walking and continuously provides information to the nervous system about the characteristics of the walking surface. Any change in the support surface thus leads to the alteration of the information from the sensory afferent of the foot and results in the modification of the loading pattern of foot and other kinematic changes (Eils et al 2004). In our study, the significant changes in the spatio-temporal gait parameters suggest that the experimental surfaces did create a challenging environment in terms of modification of gait patterns. Alterations in the plantar pressure parameters support the hypothesis that the simulated outdoor surfaces did produce significant changes in the foot to surface interaction. The results of the correlation analysis between ankle angle and the centre of mass indicated that subjects were less stable on the experimental surfaces as compared to the control surface. There was no significant change observed in the cross correlation value for the trunk - shank acceleration signals and in the centre of mass for different surfaces which indicated that the disturbances were absorbed at the level of the foot and were not transmitted to the upper body. The simulated outdoor surfaces in this study, proved to be effective in creating small disturbances and challenging balance control during the steady state of walking.

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