AN ABSTRACT OF THE DISSERTATION OF

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Although research evidence clearly indicates support surface properties are a major factor contributing to fall risk among the elderly, investigations examining the influence of variations in surface conditions on the postural control of older adults during task performance have been limited. Thus, the primary purpose of the present studies was to determine whether the coordination and control of body kinematics exhibited by older adults during upright leaning (i.e., leaning forward through the region of stability) and gait would be different across variations in support surface properties. Secondary objectives of these studies included, examining if coordination and control measures of body kinematics differed as a function of the participants' level of postural stability and/or repeated exposure to the support surface properties. Three support surface conditions were selected for inclusion based on resistance properties to applied forces (i.e., normal and shear): rigid, high friction; compliant; and rigid, low friction. For both tasks performed, body kinematics for trials 1-3 (T1) and 10-12 (T2) from 12 completed trials on each support surface were analyzed using three-dimensional (3-D) video analyses. Results of separate univariate repeated measures analyses of variance yielded significant surface condition main effects for lower extremity coordination patterns and postural control strategies in the gait and leaning task, respectively. Additionally, a significant surface condition main effect and an interaction effect of surface condition by trial block were identified for the measure of head stability in the gait and leaning tasks, respectively. Differences in head stability and the control of lower extremity joint motions as a function of level of postural stability (i.e., group differences) were observed only during the walking task. Present findings indicate that during goal-directed

behavior, the coordinated movements of the body and its segments emerge from constraints imposed by the interaction of the support surface, the task and the individual. The observed adaptations in the coordination and control of posture in response to support surface constraints evidenced in the present studies provide support for the theory of perception and the control of bodily orientation (Riccio & Stoffregen, 1988).

Task and Support Surface Constraints on the Coordination and Control of Posture in Older Adults

by

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I understand that my dissertation will become part of the permanent collection of Oregon State University libraries. My signature below authorizes the release of my dissertation to any reader upon request.

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Sean Clark, Author

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INTRODUCTION

Many tasks and activities associated with daily living require control of upright posture on a wide variety of support surfaces (Tideiksaar, 1990). For example, daily routines within one's home environment often consist of standing on and/or walking over different support surfaces, such as thick pile carpet, highly polished or wet linoleum, and/or slip resistant ceramic tiles. Control of posture on any specific surface, however, is limited or constrained by the properties of that surface (e.g., friction, firmness, extent, and/or inclination) (Horak & Nashner, 1986; Stoffregen & Flynn, 1994). That is, the coordinated movements of the body and its segments effective for action on one surface may not be appropriate for a surface with different support properties. For example, people walk differently on a slippery linoleum floor as compared to a thick pile carpet. The ability to adapt the control of posture to variations in the support surface properties is necessary to maintain postural stability and minimize the risk for falls (Patla, Frank, & Winter, 1990). Consequently, the successful performance of many everyday tasks requires that a person is not only able to pick-up the relevant information about support surface properties for the control of action, but also adapt the control of posture to changing surface constraints. (Riccio & Stoffregen, 1991; Stoffregen & Flynn, 1994).

Performance of many daily activities within the home and community often places older adults at greater risk for loss of balance and/or falls (Tideiksaar, 1989). Control of posture while one stands and/or maneuvers about within the home environment becomes increasingly difficult due to age-associated changes in the perceptual (e.g., visual, vestibular, and somatosensory) and action (e.g., musculo-skeletal) systems (Cutson, 1994; Tinetti, Speechley, & Ginter, 1988). These age-associated changes in the perceptual and action systems influence a person's ability to identify and appropriately adapt the control of posture to variations in the support surfaces properties (Horak, Shupert, & Mirka, 1989). Consequently, many elderly persons are at moderate or high risk for falls when performing tasks that require them to

stand and/or walk on slippery, compliant and/or narrow surfaces (Gabell, Simmons, & Nayak, 1985). Thus, to better understand postural control as it relates to falls and fall risk among the elderly, researchers and practitioners should examine movements and actions as individuals interact with the environment (especially different support surface properties) while performing goal-directed behavior(s).

Review of Literature

Maintenance of upright posture is an active process whereby the interaction between a person and the environment is controlled through the pickup of information by the perceptual systems (e.g., visual, vestibular, and somatosensory) and the stable, efficient, coordination of movement across multiple body segments (Riccio & Stoffregen, 1988; Riccio & Stoffregen, 1991). One group of researchers (i.e., Riccio & Stoffregen, 1988; Stoffregen & Riccio,1988; Stoffregen & Flynn,1994) has recently postulated that the information necessary to appropriately perceive the dynamics of the environment (e.g., support surface properties) is "specified" through the kinematic stimulation of the visual, vestibular, and somatosensory systems (i.e., intermodal stimulation). Moreover, the patterns of stimulation across the perceptual systems are invariant or lawfully related to the interaction between an individual and the environment (i.e., intermodal invariants) and thus provide specific information for the control of action (Stoffregen & Riccio, 1988; Riccio & Stoffregen, 1991).

During goal-directed behavior, specificity between the patterns of perceptual stimulation and the dynamics of the individual-environment interaction is critical for the control of posture. As a person stands and/or walks on a support surface, the relation between information available to the visual, vestibular, and somatosensory systems (i.e., pattern of intermodal stimulation) not only indicates the degree of postural stability but also provides information regarding the utility of different movement strategies for the control of action (Riccio & Stoffregen, 1988, 1991). Consider the situation when an individual unexpectedly transitions from walking on a level surface to an inclined surface. During this transition, plantar flexion of the foot (i.e., change in the angle of the ankle joint) of the lead limb ends abruptly following heel strike due to the inclination of

the surface. In contrast, the head and trunk segments continue to progress forward. Stimulation of the perceptual systems emerging from this individual-environment interaction indicates potentially destabilizing movements of the upper-body due to the continued forward momentum (i.e., the degree of postural stability). In addition, the patterns of stimulation across the perceptual systems specify the need to both control the forward progression of the head and trunk, and alter the coordination of the lower extremities to increase the vertical trajectory of the toe of the swing limb. Failure to either appropriately perceive the dynamics of this individual-environment interaction or adapt the control of movements accordingly will result in further postural instability (i.e., a stumble or trip).

Changes in the sensitivity of the perceptual systems to stimulation (e.g., peripheral neuropathy or vestibular impairment) and/or the stimulation of the perceptual systems via intrinsic sources (e.g., postural or intentional tremor) alter the pickup of information relevant for the control of posture (Assaiante & Amblard, 1995; Horak & Macpherson, 1995; Riccio & Stoffregen, 1991). Specifically, changes in the patterns of stimulation across the perceptual systems may alter the specificity between intermodal stimulation and the individual-environment interaction (Stoffregen & Riccio, 1988; Riccio & Stoffregen, 1991). For example, for persons with peripheral neuropathy, patterns of stimulation across the perceptual systems when standing on firm, deformable (i.e., non-rigid), and narrow support surfaces may be similar due to changes in the sensitivity of somatosensory receptors in the lower extremities. Consequently, these persons not only experience difficulty maintaining upright stance on support surfaces that are deformable and/or narrow but also exhibit inappropriate postural strategies when standing on firm surfaces (i.e., use of a hip strategy) (Horak, Nashner, & Diener, 1990).

For many older adults, age-associated changes in the perceptual systems alter the patterns of intermodal stimulation specified by the individual-environment interaction (see Alexander, 1994 and Horak, Shupert, & Mirka, 1989 for review). Regardless of the temporal onset (e.g., slow or acute) of these changes, elderly persons must identify the specificity between the altered patterns of stimulation across the perceptual systems and the dynamics of the individual-environment interaction. Failure to detect this new specificity may result in an inability to correctly perceive the support surface dynamics,

including the appropriate postural strategies (Horak, 1992; Riccio & Stoffregen, 1991). Since the effective control of posture is constrained by support surface dynamics (i.e., individual-environment interaction), persons who adopt control actions "linked" to these altered patterns of stimulation may be susceptible to postural instability and an increased risk for falls (Riccio & Stoffregen , 1991; Horak, 1992). As suggested by Riccio and Stoffregen (1991), "(t)he greatest disruptions will occur when patterns of stimulation for which the (person) already has control strategies no longer correspond to situations for which those strategies are appropriate"(p. 211).

Alterations in the patterns of stimulation across the perceptual systems may also be attributed to instability of the perceptual systems (Riccio & Stoffregen, 1991). That is, uncontrolled movements of segments (i.e., the perceptual systems) give rise to patterns of stimulation that are independent of the surface dynamics (Stoffregen & Riccio, 1988; Assaiante & Amblard, 1995). As a result, postural stability may be compromised whenever a person adapts the control of posture "to patterns of stimulation that have ceased to be specific to those environmental conditions for which the control is appropriate"(Riccio & Stoffregen,1991; p.212). For example, self- or internally-generated motions of the head (i.e., stimulation of the visual and vestibular systems) may preclude detection of external perturbations and/or the dynamic properties of the environment (Riccio & Stoffregen,1988, 1991; Holt, Jeng, Ratcliffe, & Hamill,1995). Thus, stability of the visual and vestibular systems is crucial for the pickup of relevant information necessary for action (e.g., the control of posture).

Head stability may be a constraint on upright posture that emerges when visual and/or vestibular information is important for achievement of goal-desired action. Research has demonstrated that head stability is often maximized when individuals attempt to maintain an upright posture during difficult postural activities (Assaiante & Amblard, 1990; Lekhel, Marchand, Assaiante, Cremieux, & Amblard, 1995; Pozzo, Berthoz, & Lefort, 1990; Pozzo, Berthoz, Lefort, & Vitte, 1991; Pozzo, Levik, & Berthoz, 1992). Regardless of whether the task demands static or dynamic postural control (e.g., bending to pick up an object from the floor or standing on a bus), individuals will seek to minimize motions of the head induced by internal and/or external perturbations when the pickup of visual and vestibular information is critical for

successful task performance (Holt et al., 1995). "It follows that an essential characteristic of behavior is the effective maintenance of the orientation and stability of the sensory and motor platforms (e.g., head and shoulders) over variations in the (person), the environment, and the task" (Riccio, 1993; p. 321).

Maintenance of head stability during the performance of various postural tasks may be achieved through flexibility in joint and/or muscular actions (Holt et al., 1995; Pozzo et al., 1992; Winter, MacKinnon, Ruder, & Wieman, 1993). Since the human body is a multi-link system, variability in the temporal and/or spatial patterning of movements across body segments acts to filter or attenuate perturbations (e.g., impact forces associated with heel strike) that may produce head instability (Holt et al., 1995; Winter, Patla, Frank & Walt, 1990). Specifically, during both stance and gait, regulation of lower extremity muscular activity serves to minimize joint stiffness and thereby promotes the "unfreezing" of degrees of freedom and the subsequent reduction in the phase-locking of segments (Holt et al., 1995; Riccio & Stoffregen, 1991; Winter, 1989). Indeed, several investigators have identified variability in the relative phasing of joints to be crucial for the control of erect posture and head stability during gait (Holt et al., 1995; Winter, 1989).

The coordinated stabilization of body segments not only "... minimize(s) motions that degrade or interfere with perception and action (Riccio & Stoffregen, 1991; p. 199)" but also constrains postures or body configurations. Maintenance of upright posture involves continuous control and coordination of movement across one or more of the body's joints (e.g., hip and ankles). Variation in the kinematic (i.e., temporal and spatial) patterns across these joints allows for the movement of multiple body segments to be coordinated in a variety of ways (Nashner, 1989; Riccio & Stoffregen, 1988). Consequently, various patterns of coordination, or strategies, may be used to adjust the position of the center of mass (COM) to maintain an upright posture. Differences in the patterns of coordination for the control of anterior-posterior COM sway have, in fact, been demonstrated. Nashner and colleagues (Nashner & McCollum, 1985; Horak & Nashner, 1986), using a perturbation paradigm (i.e., translation and rotation of the support surface), have identified three primary automatic postural reactions utilized in the control of unexpected A/P displacements of the COM. These compensatory postural

strategies include in their purest form, an ankle strategy, a hip strategy and a stepping strategy. The relative effectiveness of each of these strategies for the control of upright posture is, however, constrained by the support surface on which the task is performed (Nashner & McCollum, 1985; Horak & Nashner, 1986; Stoffregen & Flynn, 1994).

Although a variety of postural strategies may be adopted on a given support surface, Riccio and Stoffregen (1988,1991) suggest that the dynamics of the individual-environment interaction limit or constrain the effective control and coordination of body segments during upright stance and gait. Variations in the properties of the support surface (i.e., firmness, friction, extent, and inclination) alter the dynamics of this individual-environment interaction and thereby influence the control of upright posture (Riccio & Stoffregen, 1988, 1991). For example, variations in the support surface extent (e.g., narrow or broad surface relative to the feet) influence the effectiveness of different postural control strategies (pattern of coordination and control of body segments; Riccio & Stoffregen, 1988) for the maintenance of upright stance (Horak & Nashner, 1986). That is, the anterior-posterior body sway during stance is controlled primarily by forces applied to the support surface due to torques generated at the ankle joints. However, these same ankle torques on a conforming or non-rigid surface, may cause the foot to depress into the surface and thereby reduce the effectiveness of this postural strategy for maintaining control of upper body movements (Riccio & Stoffregen, 1988). Empirical support for the influence of surface constraints on the control of posture has been demonstrated by Horak and colleagues (Horak, 1992; Horak, Shupert, & Mirka, 1989). Using perturbation paradigms (e.g., support surface rotation and translation), these investigators demonstrated that the effectiveness of postural strategies in response to postural perturbations was influenced by the size of both the support surface and the perturbation.

In regards to the constraints on postural control imposed by the support surface properties, the performer must both be attuned to these constraints and possess the abilities to achieve the goal-desired action (e.g., control of upright posture) in light of the constraints (Riccio & Stoffregen, 1991). For many persons, movements can be adapted almost immediately and unknowingly to match a changing environment. That is, individuals are able to quickly differentiate differences in the properties of the surface

and appropriately adjust the control of posture to these new support surface constraints (Strandberg, 1983; Patla, Winter, & Frank, 1990; Riccio & Stoffregen 1988). For example, when repeatedly walking across a slippery surface individuals decrease the angle of the foot relative to the support surface to minimize the shear forces associated with each heel strike (Andres & O'Connor, 1990). Alterations in the angle of the foot at heel strike serves to reduce the risk of slipping (Andres & O'Connor, 1990). From this research example, it is evident that persons are generally able to perceive the surface dynamics quickly and make necessary adaptations to control of posture.

On a novel surface, however, the dynamics of the individual-environment interaction may not be immediately "discovered". A failure to adapt the control of posture to the new individual-environment dynamics will often result in postural instability. Riccio and Stoffregen (1991) suggest, "In this condition the (person) is functionally insensitive to the interaction between the new dynamics and its own performance. That is, the dynamics are not perceived in terms of their consequences for performance" (p.204). If this instability is severe, loss of balance and/or falls may occur. If however, the instability is less severe, the older adult may either cease engagement in the activity, vary or relax the goals associated with the behavior, or remain in a prolonged state of postural instability while further exploring the underlying individual-environment dynamics (Riccio & Stoffregen, 1991).

Purpose

Adaptive control of posture in response to altered support surface dynamics generally occurs unnoticed until this type of control is compromised. For many older adults, especially those at high risk for falls, age-associated changes in the perceptual (e.g., visual, vestibular, and somatosensory) and action (e.g., musculo-skeletal) systems, as well as declines in anticipatory actions make it difficult to adapt the control of posture to variations in the support surface dynamics. Such age-associated declines in the perceptual and action systems may contribute to an older adult's inability to correctly perceive the dynamics of the support surface, and/or "... assemble and execute the control actions that are appropriate for the new dynamics"(Riccio & Stoffregen, 1991; p.204).

That is, although the available patterns of stimulation across the perceptual systems specify the dynamics of the individual-environment interaction (i.e., the possible control actions), persons may fail to pickup the information relevant to the control of posture. Indeed, Horak and colleagues (Horak, 1992; Horak, Shupert, & Mirka, 1989) have demonstrated that older adults with diagnosed balance-related disorders (e.g., changes in somatosensory and/or vestibular sensitivity) often utilize postural strategies that are ineffective given the constraints of the support surface properties.

For many older adults, the interaction between support surface dynamics and age-associated changes in the perception and action systems may predispose them to falling. Specifically, some older adults may be unable to perceive the constraints of the surface and/or appropriately adapt the coordination and control of posture (i.e., alter joint kinematics) to these constraints. For these persons, postural stability is compromised and their overall risk for falls increases significantly. The purpose of these investigations is to examine the coordination and control of posture during both stance and gait across a variety of support surfaces.

Chapter 2

SUPPORT SURFACE CONSTRAINTS ON THE COORDINATION AND CONTROL OF STANCE IN OLDER ADULTS

Sean Clark

Abstract

A major premise underlying the theory of perception and control of bodily orientation (Riccio & Stoffregen, 1988) is that the control and coordination of body segments during goal-directed action is constrained by the dynamic properties of the support surface. The present investigation examined support surface constraints on the perception and control of bodily orientation by determining the postural control strategies and degree of head stability exhibited by older adults (OAs) when leaning forward maximally through their stability region on each of three surfaces. Eight healthy OAs and 8 OAs identified at-high risk for falls completed 12 forward leaning trials on each of three support surfaces: rigid, high-friction, compliant, and rigid, low-friction. Body kinematics for trials 1-3 (T1) and 10-12 (T2) from the 12 completed trials for each surface condition were analyzed using three-dimensional (3-D) video analysis. Cross-correlation analyses were performed using trunk and leg angular position time-series data from each trial. The calculated cross-correlation coefficients were used to identify postural control strategies adopted by participants on each surface. The average Root Mean Square (RMS) accelerations of head motions in the sagittal plane were also calculated for T1 and T2. Results of separate univariate 2 x 3 x 2 (Group x Surface Condition x Trial Block) RM ANOVAs yielded a significant Surface Condition main effect for cross-correlation values ($p \leq .001$) and a significant Surface Condition by Trial Block interaction for RMS head acceleration ($p \leq .007$). Scheffe's post-hoc comparisons further indicated that the postural strategy adopted on the compliant surface was significantly different than that exhibited on the rigid, high-friction surface. These findings indicated that the coordinated movements of the body and its segments effective for action on one surface were not appropriate for certain other surfaces with different support properties. Additionally, RMS head acceleration values for the compliant surface were significantly higher than the other support surfaces for T1 only. Following repeated exposure to compliant surface dynamics, the degree of head stability during task performance increased irrespective of the participant's level of postural stability. In conclusion, the collective findings from the present investigation provide empirical support for the theory of perception and control of bodily orientation.

Introduction

Falls and fall-related injuries are a common and serious concern for many elderly persons. In fact, research evidence indicates that approximately 1/3 of community-dwelling older adults over the age of 65 years fall at least once each year (Campbell, Borrie & Spears, 1988). Of the elderly persons who fall, approximately 5-15% suffer severe physical injuries (e.g., bone fractures, sprains) that often result in nursing home admission and/or long-term care (Nevitt, Cummings, Kidd, & Black, 1989; Rubenstein, Robbins, Schulman, Rosado, Osterweil, & Josephson, 1988; Tinetti, Speechley, & Ginter, 1988). The repercussions of non-injurious falls are often equally devastating. For many older adults, non-injurious falls result in psychological trauma or an increased "fear" of falling (Tinetti & Speechley, 1989). This heightened fear of falling, in turn, leads to a decline in functional autonomy evidenced by the self-imposed discontinuation of many physical activities, disengagement from social involvement, and dependence on others for the performance of daily tasks (Cutson, 1994; Rai & Kiniorns, 1995; Tinetti & Speechley, 1989).

The etiology of falls is multifactorial in nature (Nevitt, Cumming, & Hudes, 1991; Prudham & Evans, 1981; Rubenstein & Robbins, 1990; Tinetti & Speechley, 1989). Indeed, research evidence indicates that the increased risk for falls among older adults may be attributed to factors that are; (a) intrinsic to the individual, (b) associated with the nature of the activities or tasks performed, and/or (c) related to the properties of the environment (Cutson, 1994; Rubenstein et al., 1988; Tinetti & Speechley, 1989). Although researchers and clinicians widely recognize individual, task, and environmental factors as contributors to fall-risk, the complex interaction by which these factors predispose a person to falls is rarely studied. Instead, research investigations generally examine the independent contributions of one or more of these factors on fall-risk for older adults (Whipple, Wolfson, & Amerman, 1987; Overstall, Exton-Smith, Imms & Johnson, 1977). Although this research provides important insights into how system impairments and/or environmental hazards impact functional performance (e.g., balance and mobility), this approach neglects to explore the interrelationship of individual, task, and environmental factors as they influence fall-risk concurrently. Consequently, the risk for falls is often determined from the cumulative number of factors identified that

negatively impact balance and/or mobility as opposed to how the interaction of these factors influences functional performance (Mayo, Korner-Bitensky & Levy, 1993; Tinetti, Williams & Mayewski, 1986; Woolley, Czaja & Drury, 1997).

One theoretical perspective that emphasizes the interaction of individual, task, and environmental constraints on the control of posture and falls is the theory of perception and the control of bodily orientation recently developed by Riccio and Stoffregen (1988). Within this theoretical framework, the interactions between the person, environment, and task are considered fundamental to the control and coordination of posture and movement (Riccio & Stoffregen, 1988; Stoffregen & Riccio, 1988). Specifically, the maintenance of postural control is viewed as an active process whereby the interaction between an individual and the environment is controlled through the pick-up of information in the environment via the perceptual systems (e.g., visual, vestibular, and somatosensory), and the stable, efficient coordination of movement across multiple body segments to achieve a goal-desired action (Riccio & Stoffregen, 1988, 1991). From this theoretical perspective, research on postural control as it relates to falls among the elderly, should emphasize the study of movements and actions as individuals interact with the environment (e.g., support surfaces) while performing goal-directed behavior(s).

Support surface constraints and adaptive postural control

Although support surface properties have been identified as an environmental factor that contributes to the increased risk for falls among older adults, only a limited number of studies have examined exactly how surface properties constrain the control of posture (Alexander, Shepard, Gu, & Schultz, 1992; Marin, Bardy, Baumberger, Fluckiger, & Stoffregen, 1997; Lekhel, Marchand, Assaiante, Cremieux & Amblard, 1994; Teasdale, Stelmach, & Breunig, 1991; Wade, Lindquist, Taylor, & Treat-Jacobson, 1995). Of these investigations, most have examined postural control in response to an externally imposed threat to balance. For example, postural responses to visual illusions of self-movement and support surface translations and/or rotations have been assessed while persons attempt to maintain an upright stance on either a rigid, compliant (foam) or narrow (beam) support surface (Alexander et al., 1992; Teasdale, Stelmach, & Breunig, 1991; Wade et al., 1995). The collective evidence emerging from these investigations

indicates age-associated declines in compensatory postural control in response to perturbations during upright stance. Despite the extensive research efforts in which perturbation paradigms have been used to assess fall-risk, research evidence indicates that postural control may be quite different when comparing volitional movement to that resulting from externally-imposed perturbations (DiFabio & Emasithi, 1997; Marin et al., 1997; Patla, Frank, & Winter, 1990; Stoffregen, Adolph, Thelen, Gorday, & Sheng, 1997). These findings suggest the need to further explore the interaction between individual and support surface constraints on the control of posture during unperturbed stance.

Riccio and Stoffregen (1988, 1991) have recently suggested, that during unperturbed upright stance, the effective control and coordination of body segments is constrained by dynamics of the individual-environment interaction. Specifically, postural configurations and the coordination of body segments during stance are influenced by the dynamic and static-structural properties of the support surface on which one stands (e.g., firmness, extent, inclination, friction) (Stoffregen & Flynn, 1994; Stoffregen & Riccio, 1988). For example, on a rigid, narrow support surface, anterior-posterior postural sway may be effectively controlled by muscular action at the hip joint (i.e., use of a hip strategy) (Horak & Nashner, 1986; Nashner & McCollum, 1995). Movements associated with this hip strategy generate shear forces at the individual-support surface contact region and thus allow for the effective control of posture. However, if these same hip movements were replicated on a low friction surface, the shear forces applied to the slippery surface might cause the feet to slip or slide and thereby reduce the effectiveness of this postural strategy.

Head stabilization and postural control

Critical to the maintenance of upright posture is the integrity of the perceptual information that specifies the appropriate action for a given support surface (Riccio & Stoffregen, 1988, 1991). That is, to effectively control and coordinate posture while standing and/or walking on a surface, the information made available from the perceptual systems must enable the constraints of the support surface to be appropriately identified (Patla et al., 1990; Riccio & Stoffregen, 1991). A number of investigators have recently

suggested that perception of support surface properties and/or external perturbations is influenced by the relative stability of the visual, vestibular and/or somatosensory systems (Horak & Macpherson, 1995; Holt, Jeng, Ratcliffe, & Hamill, 1995; Keshner & Chen, 1996; Riccio & Stoffregen, 1991). Thus, declines in head stabilization and the associated instability of the visual and vestibular systems may compromise the integrity of the perceptual information requisite for the detection of support surface properties (Holt et al., 1995; Pozzo, Levik, & Berthoz, 1992). Increased stability of head trajectories, therefore, may be critical when the pickup of visual and/or vestibular information is necessary for the achievement of goal-directed actions. Indeed, the reported findings from several investigations have indicated that head stability was maximized when young adults were required to maintain an upright stance during difficult postural tasks (e.g., standing on foam, unilateral stance on a narrow beam) (Assiante & Amblard, 1990; Lekhel et al., 1994; Pozzo, Levik, & Berthoz, 1992).

DiFabio and Emasithi (1997) have recently demonstrated that older adults also maximized head stability when performing difficult postural tasks. These authors report that older adults, with no reported vestibular symptoms, minimized head motions to a greater extent than young adults when upright stance was coupled with sway-driven tilt of the support surface and/or visual surround. These preliminary investigations further indicated that when performing postural tasks in which perceptual inputs were compromised, older adults effectively increased the stability of head trajectories by altering coordination strategies associated with head motions (i.e., adopted a head-stabilization-in-space strategy). DiFabio and Emasithi (1997) suggested that the older adults in these investigations might have increased the stability of head trajectories during the voluntary, self-initiated postural tasks to compensate for potential age-associated declines in the perceptual systems. That is, the increase in head stabilization exhibited by older adult participants served to optimize the integrity of the visual and vestibular inputs associated with head motions and thereby facilitated the appropriate control of head and trunk orientation necessary for the maintenance of upright stance. DiFabio and Emasisthi (1997) further postulated that stability of head motions is critical for individuals who experience difficulties integrating perceptual information and/or for the successful

performance of difficult postural tasks (e.g., stance on compliant surface, walking in a dimly lit room).

Postural control in older adults

For many older adults, especially those at-high risk for falls, involvement in activities that demand adaptive control of posture to variations in support surface properties often has negative consequences for goal-related behavior (Tideiksaar, 1990). Age-associated declines in the perceptual (e.g., visual, vestibular, and somatosensory) and action (e.g., musculoskeletal) systems often make it difficult for older adults to correctly perceive the properties of the support surface and to employ the appropriate control strategy for these new constraints (Horak, Shupert, & Mirka, 1989). Failure to appropriately identify and adapt the control of posture to the new individual-environment dynamics will often result in postural instability. Riccio and Stoffregen (1991) suggest, "In this condition the (person) is functionally insensitive to the interaction between the new dynamics and (his/her) own performance. That is, the dynamics are not perceived in terms of their consequences for performance" (p.204). If this instability is severe, loss of balance and/or falls may occur. If, however, the instability is less severe, the older adult may either cease engagement in the activity, vary or relax the goals associated with the behavior (e.g., use assistance for support), or remain in a prolonged state of postural instability (Riccio & Stoffregen, 1991). Attempts to maintain upright posture during prolonged states of instability often provide persons with opportunities to further explore the underlying individual-environment dynamics (Riccio & Stoffregen, 1991). Consequently, this additional exploratory activity may ultimately result in the perception of the individual-environment dynamics and the effective coordination and control of posture (Horak & Nashner, 1986).

Uncertainty about one's abilities to adapt the control of posture to variations in the support surface dynamics, as well as a knowledge of the negative consequences associated with failure to make appropriate postural adaptations often results in a reduction in the level and variety of activities in which an older adult is willing to participate (Lawton & Brody, 1969; Rai & Kiniorns, 1995; Tinetti & Speechley, 1989). Despite the potential for such lifestyle changes, van Weel, Vermeulen and van den Bosch

(1995) suggest that, "(m)ost falls are related to movements, and since elderly people will want to preserve their freedom to move around, slipping and tripping cannot be avoided altogether" (p.1550). Although research evidence has clearly indicated that environmental factors, especially surface conditions, contribute to the increased risk for falls among community-dwelling older adults, a detailed investigation that examines how support surface properties constrain postural control (i.e., body kinematics and head stability) during upright stance across a variety of support surface properties has yet to be conducted.

Study objectives and hypotheses

The purpose of this first study was to examine the influence of variations in support surface dynamics on the coordination and control of upright posture when older adults leaned forward maximally through their stability region. Specifically, the study had four objectives: (A) examine whether postural coordination (i.e., postural control strategies) would be different across variations in support surface dynamics, (B) contrast adaptations in postural coordination following repeated exposure to surface dynamics for that of older adults identified at-high risk for falls (i.e., at-high risk for falls group) with that of healthy older adults with no history of falls or postural instability (i.e., healthy group), (C) determine whether head stability would be different across variations in support surface dynamics, and (D) contrast the degree of head stability exhibited by the at-high risk for falls group with the healthy older adult group. It was hypothesized that: (A) older adults would utilize surface-appropriate postural coordination strategies, (B) healthy older adults, in contrast to older adults at-high risk for falls, would demonstrate appropriate adaptations in postural coordination following repeated exposure to support surface dynamics, (C) head stability irrespective of level of postural stability would be maximized when upright posture was maintained on compliant and low-friction surfaces, and (D) the degree of head stability evident in the at-high risk for falls group would be significantly greater than that observed for healthy older adults across all surface conditions.

Methods

Participants

Eight healthy older adults and eight older adults (76-86 years) identified at-high risk for falls, were recruited from the community to participate in the present study. These persons were identified from a pre-study screening that provided a comprehensive clinical assessment of balance and mobility. Assessments in the pre-study screening included: (a) a background and medical history questionnaire (see Appendix A), (b) a Berg Balance Scale (Berg, 1993) (see Appendix B), (c) a Limits of Stability Test® (LOS) conducted at 100% of the participant's theoretical limits of stability, and (d) a Sensory Organization Test® (SOT). Prior to participation, all persons provided written informed consent in accordance with policies outlined by the University Institutional Review Board at Oregon State University (see Appendix C).

From the results of 55 older adults who completed the pre-study screening tests, eight persons met the specific eligibility criteria for the healthy group and nine individuals met criteria for the group identified at-high risk for falls (one individual in the at-high risk for falls group did not participate in the study due to an unexpected illness). Eligibility criteria for the healthy older adult group were as follows: (a) living independently in the community (i.e., non-institutionalized setting), (b) no persistent problems with dizziness or unsteadiness, (c) no musculoskeletal impairments, (d) a performance score greater than 50/56 on the Berg Balance Scale, (e) a mean Maximum Excursion value on the Limits of Stability Test greater than 80% and, (f) an equilibrium performance score on the SOT greater than 70. In contrast, criteria for the group identified as being at-high risk for falls were as follows: (a) living independently in the community (i.e., non-institutionalized setting), (b) a self-report of postural instability and/or repeated falls in the previous 2 years, (c) performance scores equal to or below 50/56 on the Berg Balance Scale, (d) a mean Maximum Excursion value on the LOS test less than 80% and, (f) an equilibrium performance score on the SOT equal to or less than 70. Persons with diseases known to adversely affect balance and mobility (e.g., Parkinson's disease, multiple sclerosis, CVA, diabetes), permanent orthopedic

impairments (e.g., fused joints, amputation) and/or visual deficits not correctable with lenses were excluded from participation in the study.

The criteria scores used to establish group eligibility were determined from pilot study data and previously reported performance scores for the Berg Balance Scale (Rose & Clark, 1995). Specifically, pilot data from the performance scores of 144 community-dwelling older adults (with and without a self-reported history of falls) between the ages of 65 and 92 years were used to calculate means and standard deviations for the SOT and LOS tests. The calculated mean equilibrium score of 70 and mean Maximum Excursion score (i.e., composite score from the 8 targets) of 80% LOS, from these pilot data, were used as the eligibility cut-off scores for the SOT and LOS tests, respectively. In addition, the criteria score of 50/56 for the Berg Balance Scale was selected based on previously reported performance scores from a group of older adults at-risk for falls (Rose & Clark, 1995).

Table 1.1. Group eligibility criteria data.

	Older adults at-high risk for falls	Healthy older adults
Number (N)	8	8
Male	3	4
Female	5	4
Mean Age (years)	80.9	80.6
Age Range (years)	76-86	77-83
Berg Score	48 (2) *	54 (2)
Maximum Excursion (% LOS)	69.6 (9.5) *	87.5 (4.3)
Equilibrium Score (% Stability)	63.9 (6.4) *	79.5 (4.8)

Performance scores for older adults at-high risk for falls compared to healthy older adults (* significant at the $p \leq .001$ level).

Means and standard deviations for the group criteria data for the two groups of older adults in the present investigation are presented in Table 1.1. One-way analysis of variance (ANOVA) procedures were performed separately for each criteria measure to ensure that the two groups were significantly different with respect to each of the three

performance measures ($p \leq .001$). In addition, a one-way ANOVA was performed for the dependent variable age to confirm that the two groups did not differ with respect to age.

Equipment

Support surfaces

Postural control strategies were evaluated while each participant leaned forward maximally through her/his stability region on each of three different support surfaces: (A) firm, high-friction surface, (rubber matting), (B) compliant surface, (open cell, polyurethane foam) and, (C) firm, low-friction surface, (vinyl flooring coated with a thin film of glycerin-water solution). Test surfaces were selected based on their resistive properties to both torques and shear forces. The firm, high-friction surface was resistive to both ankle torques and shear forces, and thus was able to support both ankle and hip postural control strategies. In contrast, the compliant surface had low resistance to ankle torques and the firm, low-friction surface had low resistance to shear forces. Consequently, the resistive properties of the compliant and firm, low-friction surface reduced the effectiveness of ankle and hip postural control strategies, respectively. Prior to testing, each test surface was secured to the laboratory floor to prevent shifting of the surface.

Kinematics

A standard link-segment model for the foot, shank, thigh, trunk, and head was developed to assist in the kinematic analyses of coordination patterns (i.e., postural control strategies) utilized by participants while leaning forward maximally through their stability region on each of the three support surfaces. Using double-sided adhesive tape, retroreflective spherical markers (15mm) were secured to bony landmarks on the right side of the body corresponding to the fifth metatarsal (toe), lateral portion of the calcaneous (heel), lateral malleolus (ankle), lateral femoral epicondyle (knee), greater trochanter (hip), greater tubercle of the humerus (shoulder), and temporal process of the mandible (head). Participants were provided with lycra bike shorts and a tank top shirt to wear during all testing procedures.

The coordination of the body segment motions during the forward leans on each of the three support surfaces was analyzed using three-dimensional (3-D) video analyses. To attain maximum accuracy of the marker trajectories and thereby reduce potential digitizing error, each camera's field width was minimized to produce a large video image size. The 3-D motions of the reflective markers during each test trial were recorded using two synchronized video cameras (Panasonic DT5100) and external video tape recorders. The cameras were configured with an exposure time of 0.001 seconds and a frame rate of 30 Hz. The cameras were positioned 5.7 meters and 5.4 meters from the middle of a six-meter walkway with an angle of 95 degrees between the cameras' optical axes (see Figure 1.1 for experimental setup). This camera placement provided an unobstructed right sagittal view of each participant as she/he leaned forward through her/his stability limits on each surface. Prior to data collections, a calibration structure was positioned in the middle of the walkway and video recordings were obtained. Digitized data of the calibration structure were used in subsequent calculations to calibrate the measurement volume.

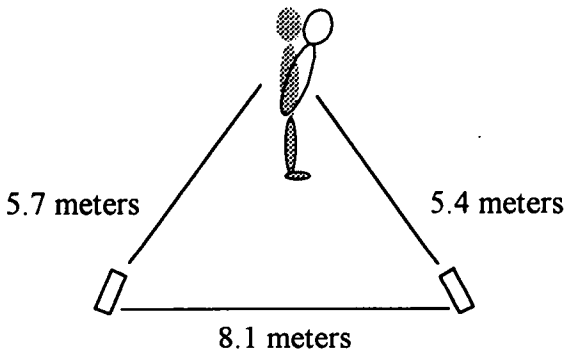

Figure 1.1. Experimental set-up.

Procedures

For safety, each participant was secured to an overhead harness system prior to the start of testing. The overhead harness was designed to allow participants to stand unassisted and lean unrestricted through their stability limits, yet prevent the occurrence

of a fall. In addition, assistance was provided to the participant as she/he stepped onto the test surface. Once the participant was able to stand independently on the surface, she/he was instructed to stand with arms extended and hands resting on the front of the thighs. The participant was then asked to look straight-ahead and lean forward out over the feet (i.e., base of support) as far as possible without losing balance or moving her/his feet. An assistant stood in close proximity behind and to the left of the participant during testing to offer support when required and/or specified by the participant. Each participant completed 12 maximal forward leaning trials on each of the three support surface conditions. A three-minute rest period was provided to the participant prior to the start of the test trials on a new surface condition. All testing was performed with the participant standing barefoot (i.e., no shoes or socks) on each of the support surfaces. The presentation order for surface conditions was randomized for each participant.

Data Reduction

The recorded video image from each camera view for trials 1 through 3 (Trial Block I) and 10 through 12 (Trial Block II) for each of the three test surfaces was digitized at 30 Hz using the Peak Performance Motion Analysis System (software version 5.3) interfaced with a personal computer. Analyses were limited to Trial Block I (N=3) and Trial Block II (N=3) in order to obtain trials that represented the participant's initial response to the support surface properties and the participant's response after repeated exposure to the surface properties, respectively.

High frequency noise in the raw digitized coordinate data for each camera was filtered using a fourth-order, zero-phase Butterworth low-pass filter prior to calculations of the three-dimensional (3-D) coordinates. The appropriate cutoff frequency for each data point in the three dimensions was determined using the Peak Performance Optimal Filter option, which is based on the Jackson Knee Method (Jackson, 1979). Three-dimensional position-time coordinates from the two camera views were determined using the filtered coordinate data and the Direct Linear Transformation (DLT) technique (Abdel-Aziz & Karara, 1971). The 3-D coordinate calculations were used to derive kinematic parameters (i.e., angular and linear positions, velocities and accelerations) utilized in subsequent analyses of the coordination and control of upper- and lower-body

segmental motions. The velocities and accelerations values were computed using central difference formulas within the Peak Performance software.

Measures of Interest

Cross-correlation analyses have been utilized by numerous investigators to indicate the degree of coupling between body parts during movement (Lekhel et al., 1994; McDonald, van Emmerik, & Newell, 1989; Stoffregen et al., 1997; Vereijken, van Emmerik, Whiting, & Newell, 1992). For example, Stoffregen and colleagues (1997) recently used this analysis technique to examine postural strategies (i.e., ankle, hip, and mixed strategies) among toddlers when standing on different support surfaces. The present investigation also utilized cross-correlation analyses to identify the coordination patterns associated with upper- and lower-body motions when older adults leaned forward maximally through their stability regions on each of the three test surfaces. Assessment of the coordination patterns was performed using cross-correlation values derived from the angular position time-series data for the trunk (i.e., linear vector from the hip marker to the shoulder marker) and lower-body (i.e., linear vector from the ankle marker to the hip marker). Angular position data were determined as the included angle between the body segment and the projected horizontal axis. Cross-correlation analyses were used to provide an indication of whether the upper- (i.e., trunk segment) and lower-body segments were controlled dependently or independently. Specifically, if the upper- and lower-body segments moved in phase, movement of these segments were considered to be highly dependent and a strong positive correlation value was evidenced (i.e., $\simeq 1.00$). A strong, positive correlation value was indicative of the upper and lower-body segments moving as a unit (i.e., trunk and leg angles increasing together). Similarly, if upper- and lower-body angular position data were out of phase, movements of the two segments were considered to be highly dependent and a strong negative correlation value was calculated (i.e., $\simeq -1.00$). In this situation, the strong, negative correlation value was indicative of the upper- and lower-body segmental angles moving in opposite directions (i.e., shoulders moved forward and hips moved backwards). If, however, the calculated cross-correlation value was close to 0.00, then the movements of

the upper- and lower-body segments were determined to be independently controlled (Stoffregen et al., 1997).

Prior to statistical analyses, cross-correlation values were derived from the angular position data for the upper- and lower-body segments for trials 1 through 3, and trials 10 through 12 on each of the three test surface conditions. Mean trial-block correlation values (i.e., Trial Block I and Trial Block II) for each surface condition were derived following Fisher Z-transformations of the individual test trial cross-correlation values. The calculated mean trial-block correlation values were then used in all subsequent statistical analyses.

In addition, Root Mean Square (RMS) acceleration values were derived from the calculated linear acceleration profiles of the head marker in the sagittal plane (i.e., anterior-posterior direction). Prior to performing statistical analyses, measures of RMS head acceleration from test trials within Trial Block I and Trial Block II were averaged for each surface condition. The averaged RMS head acceleration values were used to provide a measure of the consistency in linear sagittal plane motions of the head segment. Consequently, averaged RMS values were used to indicate the degree of head stability while participants leaned forward through their stability regions.

Experimental Design and Statistical Analyses

A 2 x 3 x 2 (Group x Surface Condition x Trial Block) between-within design with repeated measures on the last two factors was implemented to: determine whether postural coordination and head stability would be different across variations in support surface dynamics (objectives A and C); contrast adaptations in postural coordination strategies following repeated exposure to support surface dynamics for that of older adults at-high risk for falls with that of healthy older adults (objective B); and contrast the degree of head stability for the older adults at-high risk for falls with that of healthy older adults (objective D).

Preliminary descriptive statistics were performed to determine if data for the dependent variables met the statistical assumptions associated with repeated measures multivariate analysis of variance. Results of these analyses indicated violations of singularity due to ill-conditioned and singular matrices. Consequently, separate

univariate 2 x 3 x 2 (Group x Surface Condition x Trial Block) repeated measures analysis of variance (RM ANOVA) were performed using measures for postural coordination and head stability. These analyses were performed to determine differences between groups and support surface conditions as well as differences between groups following repeated exposure to the surface dynamics (i.e., Group x Trial Blocks). For the univariate RM ANOVAs, the alpha level of significance was set at the traditional level of $p \leq .05$. Additionally, statistical significance in the surface main effect and higher order interactions were further assessed using Scheffe's post-hoc multiple comparisons.

Results

Postural Control Strategies

To determine whether postural coordination was different across variations in support surface dynamics, as well as between the two groups of older adults following repeated exposure to the support surface dynamics a 2 x 3 x 2 (Groups x Surface Condition x Trial Block) RM ANOVA was performed using derived cross-correlation coefficient values (see Table 1.2). Results of this analysis indicated a significant main effect for Surface Condition ($F(2,28) = 12.78$, $p \leq .001$) (see Figure 1.2). Scheffe post-hoc comparisons conducted for the significant Surface Condition main effect indicated that the cross-correlation coefficient values derived from the compliant surface condition were significantly larger than those for the firm, high-friction surface only. Main effects for both the Group and Trial Block factors and all higher order interactions failed to reach the level of statistical significance ($p > .05$). The RM ANOVA output table and corresponding power estimates and eta squared values are presented in Appendix D.

Significant differences in the cross-correlation coefficient values for the Surface Condition main effect suggested that participants were utilizing different postural coordination strategies across the support surface conditions. However, since cross-correlation values are measures on an ordinal scale, the results of the RM ANOVA should be interpreted with caution. Moreover, since a range of cross-correlation values is representative of a certain postural strategy (i.e., ankle, hip, or mixed ankle-hip strategy), analyses that include each cross-correlation coefficient value may not appropriately

represent the underlying postural strategies utilized during the forward lean on each support surface. Consequently, postural control strategies (i.e., ankle, hip, and mixed ankle-hip strategies) were identified from the cross-correlation coefficient data and additional qualitative analyses were performed.

Table 1.2. Mean cross-correlation coefficient values across Groups and Surface Conditions.

Surface Conditions	Older adults at-high risk for falls	Healthy older adults
Firm, high-friction	-.047 (-.98 to .97)	.322 (-.30 to .98)
Compliant	-.520 (-.97 to .84)	-.147 (-.84 to .95)
Firm, low-friction	-.257 (-.98 to .94)	.104 (-.82 to .97)

Mean cross-correlation values derived from position-time series data for trunk and leg angles relative to the projected horizontal axis. The range of cross-correlation values observed for each group of older adults across surface conditions is presented in the parentheses.

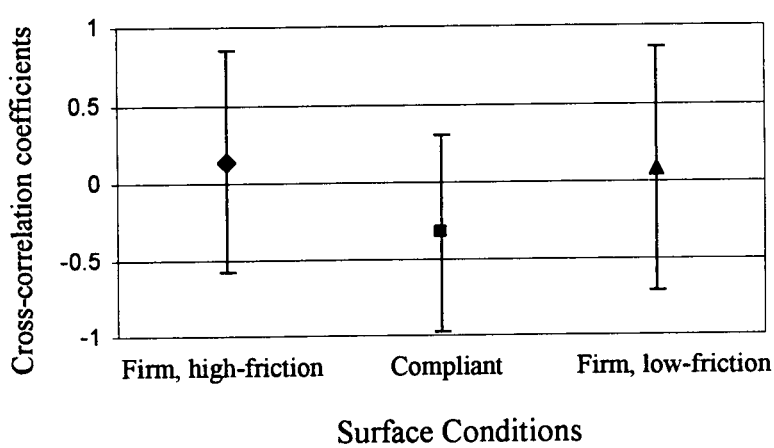

Figure 1.2. Mean cross-correlation coefficients with standard deviations across surface conditions collapsed across groups.

Postural control strategies were determined as follows: positive cross-correlation values greater than 0.50 were determined to be indicative of an ankle strategy (i.e., trunk and leg angles increasing together); a strong, negative correlation value (i.e., less than -.50) was indicative of a hip strategy (i.e., shoulders moved forward and hips moved backwards); and coefficient values between -.49 and .49 were representative of a mixed ankle-hip strategy. Since previously reported investigations have not provided criterion correlation-coefficient values for the classification of postural strategies, criteria in the present study were developed based on visual inspection of the constructed trunk-leg angle-angle diagrams. Representative trunk-leg angle-angle diagrams for ankle, hip, and mixed ankle-hip strategies are presented in Figures 1.3, 1.4, and 1.5 respectively.

To qualitatively examine differences in postural coordination across the support surface conditions, the frequency distribution of the identified postural control strategies for each support surface was determined (see Table 1.3). As indicated in this set of figures, identified hip, ankle and mixed hip-ankle strategies were observed both within and across each test surface condition. A closer examination of Figures 1.6 - 1.8 indicated that there were differences in the frequency distribution of the identified postural control strategies across support surface conditions. These findings support the results of the RM ANOVA for the cross-correlation values and indicate that participants utilized different postural strategies across variations in support surface dynamics.

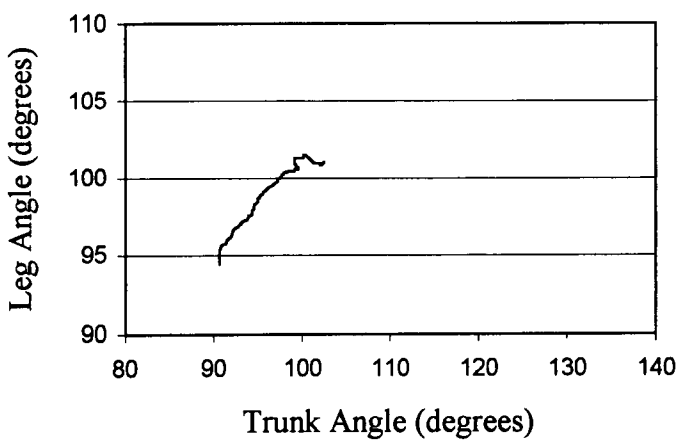

Figure 1.3. Representative trunk-leg angle-angle diagram for an ankle strategy. Cross-correlation between trunk and leg angles, $r^2 = .96$.

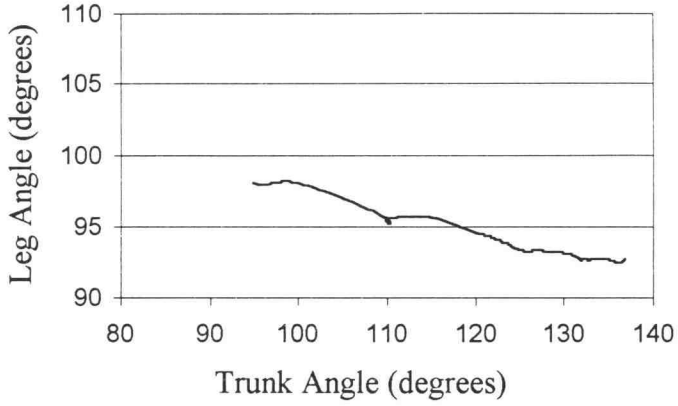

Figure 1.4. Representative trunk-leg angle-angle diagram for a hip strategy. Cross-correlation between trunk and leg angles, $r^2 = .85$.

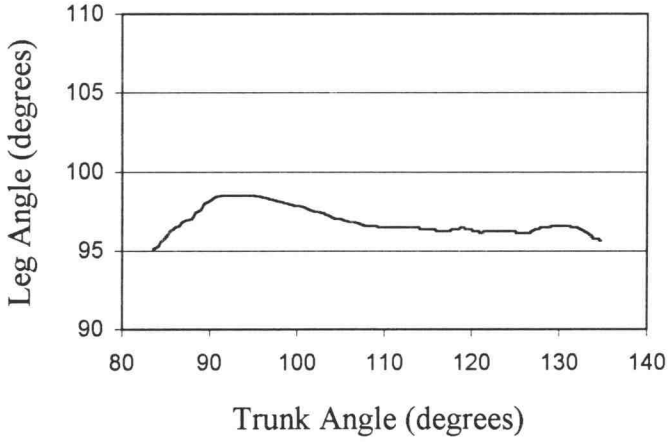

Figure 1.5. Representative trunk-leg angle-angle diagram for a mixed ankle-hip strategy. Cross-correlation between trunk and leg angles, $r^2 = .004$.

Table 1.3. Postural Control Strategy Classification across Support Surface Conditions.

	Support Surface Conditions		
Postural Control Strategy	Firm, high-friction	Compliant	Firm, low-friction
Ankle	6	2	6
Mixed ankle-hip	7	6	4
Hip	3	8	6

Values indicate the total number of participants (collapsed across groups and trial blocks) utilizing the identified postural control strategy for each support surface condition.

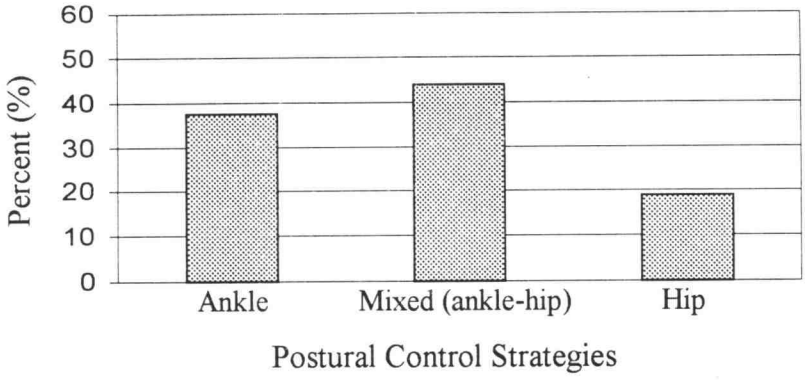

Figure 1.6. Frequency distribution of postural control strategies utilized on the firm, high-friction surface.

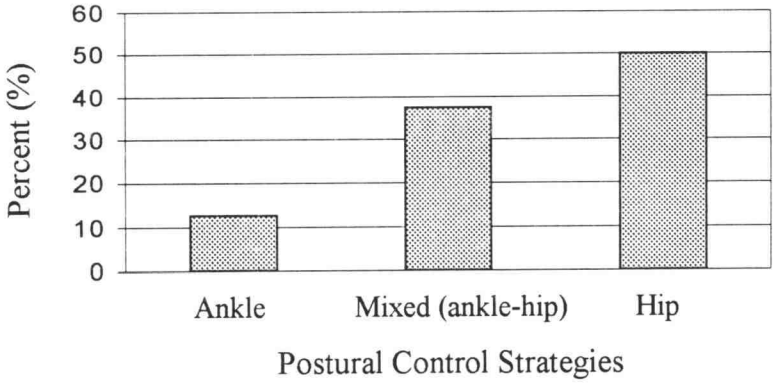

Figure 1.7. Frequency distribution of postural control strategies utilized on the compliant surface.

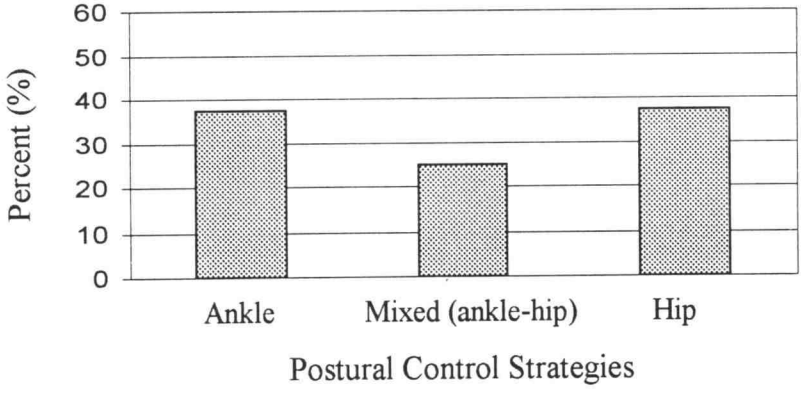

Figure 1.8. Frequency distribution of postural control strategies utilized on the firm, low-friction surface.

Head Stabilization

The mean RMS head acceleration values and corresponding standard deviations for both groups of older adults across Surface Conditions and Trial Blocks are presented in Table 1.4. A 2 x 3 x 2 (Groups x Surface Condition x Trial Block) RM ANOVA was performed using the RMS head acceleration data to examine whether head stability was different across variations in support surface dynamics and/or between the two groups of older adults. Results of the RM ANOVA indicated significant differences in mean RMS head acceleration values for the main effect of Surface Condition ($F(2,28) = 12.12$, $p \leq .001$) and the interaction of Surface Condition x Trial Block ($F(2,28) = 6.26$, $p \leq .007$). The significant Surface Condition x Trial Block interaction effect is depicted graphically in Figure 1.9. Scheffe's post-hoc multiple comparisons for the Surface x Trial Block interaction effect indicated that head acceleration values associated with performance on the compliant surface were significantly larger than head accelerations on the firm, high-friction and firm, low-friction test surfaces for Trial Block I only.

Differences in RMS head accelerations between the two groups were not determined to be statistically significant ($F(1,14) = 0.348$, $p \geq .50$). As indicated in Table 1.4, older adults at-high risk for falls tended to demonstrate a greater degree of stability in head motions as compared with healthy older adults when leaning forward through their stability regions on each test surface. The RM ANOVA output table and the corresponding power estimates and eta squared values are presented in Appendix E.

Table 1.4. Mean RMS Head Acceleration values for Groups across Surface Conditions and Trial Blocks.

		Support Surface Conditions		
Group	Trial Block	Firm, high-friction	Compliant	Firm, low-friction
At-high risk for falls	I	225.52 (93.84)	280.13 (76.58)	230.63 (60.83)
	II	223.27 (93.80)	255.11 (69.15)	241.69 (83.33)
Healthy	I	249.73 (81.22)	302.43 (91.17)	225.82 (58.16)
	II	274.25 (68.24)	282.33 (76.84)	249.86 (70.23)

Mean RMS head acceleration values (m/s^2) with standard deviations in parentheses.

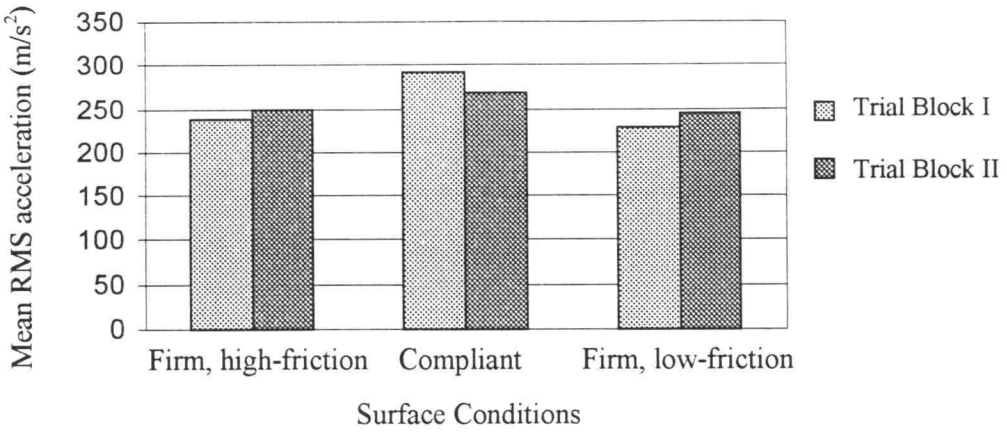

Figure 1.9. Mean RMS head acceleration (m/s^2) for the Surface Condition by Trial Block interaction collapsed across groups.

Discussion

Postural Control Strategies

The statistical analysis conducted to address the study objective of whether postural control strategies differed across variations in support surface properties yielded a significant Surface Condition main effect. This finding together with the observed differences in the frequency distribution of postural control strategies across surface conditions indicated that study participants indeed utilized different postural control strategies when performing the leaning task across variations in support surface properties. Although these results support the premise that postural control strategies emerge as a result of the imposed task and support surface constraints, the question remains as to whether the observed postural control strategies evidenced in the present investigation were appropriate for the specific surface constraints.

Closer examination of the individual participant data yielded qualitative support for the proposed hypothesis that older adults would utilize surface-appropriate postural control strategies when leaning forward through their region of stability on compliant and firm, high-friction surfaces. Since the compliant test surface had low resistance properties to ankle torques, it was expected that these surface constraints would limit the repertoire of postural control strategies effective for task performance to either a hip or mixed ankle-hip strategy (Nashner & McCollum, 1985; Riccio & Stoffregen, 1988).

Indeed, individual postural strategy data indicated that 14 of the 16 participants or approximately 85% of older adults used either a hip or mixed ankle-hip strategy when leaning forward through their stability region on the compliant surface. In contrast, only 2 of the 16 participants utilized an ankle strategy for the compliant surface condition.

Although the firm, high-friction surface was able to support both ankle torques and shear forces, researchers have suggested that when upright stance is maintained on a firm, high-friction surface the position of the COM (until the COM approaches the stability boundary) is controlled primarily by the use of an ankle strategy (Nashner & McCollum, 1985; Riccio & Stoffregen, 1988). A closer examination of individual postural strategy data in the present investigation indicated that only 3 of the 16 older adults or approximately 19% of the participants exhibited a hip strategy on the firm, high-friction surface. Collectively, the observed findings for the compliant and firm, high-friction support surface conditions indicated that older adults, irrespective of their level of postural stability, utilized surface-appropriate postural control strategies when constrained by task and support surface dynamics.

For support surface conditions to impose constraints on an observed behavior (i.e., body kinematics), the dynamic and static structural properties of the surface must have functional relevance to the task performed (Riccio & Stoffregen, 1991). That is, the properties or features of the support surface must afford and support changes in the coordinated actions of body segments necessary for the achievement of task goals. Achievement of the stated task goals in the present investigation (i.e., lean forward out over the feet as far as possible without losing balance or moving the feet), required that support surface properties were able to resist both the magnitude and type of applied forces (i.e., normal and/or shear) associated with the changes in body kinematics necessary to control anterior displacements of the COM. Thus, variations in the resistance properties (e.g., firmness and friction) of the support surfaces influenced the dynamics of the individual-environment interaction and thereby constrained the coordination and control of body kinematics for action.

Although study findings indicated that postural control strategies utilized for task performance differed significantly across certain surface conditions, examination of the individual postural strategy data indicated that older adult participants were not limited to

the use of surface-appropriate control strategies to complete the stated task goals (Marin et al., 1997; Stoffregen et al., 1997). Moreover, analyses of kinematic data indicated that older adults were able to utilize a wide variety of postural control strategies both within and across surface conditions. Specifically, coordination patterns identified as ankle, hip and mixed ankle-hip postural control strategies were evidenced both within and across each of the three test surfaces.

The use of non surface-appropriate postural control strategies by a small percentage of the study participants may be explained by the participants relaxing the task goals in order to minimize the influence of surface dynamics on postural control. During task performance, anterior displacements of the COM resulted in a non-alignment of body segments with gravitational force. This change in body kinematics generated a destabilizing torque acting to move the body's COM further away from the center of the base of support (i.e., toward the stability limits). Riley and colleagues (1997) have recently suggested that this destabilizing torque or "biomechanical instability" often conflicts with the actions or postural control strategies necessary to achieve task goals. Although the instructions in the present investigation were to lean forward as far as possible, participants were not provided with a specific achievable or overt goal (e.g., touch or reach for an object set at their maximal limits of stability). Thus, when biomechanical instability resulted from changes in body position during task performance, participants may have perceived this conflict between biomechanical stability and task goals and consequently relaxed the goals of the task in order not to lean maximally (i.e., decrease destabilizing torques). That is, individuals may have recognized the trade-off between biomechanical stability and the achievement of task goals and made the necessary adjustments in the task goals to maintain postural stability using whatever postural strategy they employed.

A second and equally cogent explanation for the lack of surface-appropriate postural strategies on the firm, low-friction surface is that the frictional properties of this surface may not have been low enough to necessitate adaptations in the coordination of posture when participants performed the maximal forward leaning task. Specifically, the magnitude of the shear forces associated with trunk rotation about the hip joint may have been inconsequential and therefore unlikely to cause the participant to slip (i.e., overcome

the coefficient of static friction). Although the firm, low-friction support surface was selected with the intention to reduce the effectiveness of hip strategies, functionally the test surface may have provided participants with adequate resistance to support a wide repertoire of postural strategies. Thus, participants may have been able to select a postural strategy that would be effective but not optimal given the intended surface constraints.

Results of the present investigation did not confirm the proposed hypothesis that healthy older adults, in contrast to older adults at-high risk for falls, would demonstrate appropriate adaptations in the postural control strategies following repeated exposure to support surface dynamics. The lack of a statistically significant Trial Block effect indicates that participants utilized similar coordination patterns to control posture across test trials within each surface condition. Data further suggest that the postural control strategies adopted by both groups of participants upon initial exposure to the surface constraints were similar to those utilized after repeated exposure to surface dynamics. Riccio (1993) has suggested that, "Body configuration or movement need not change adaptively when there is persistence of the functional topology; that is, body configuration and movement is robust or insensitive to functionally inconsequential variations in the animal, the environment, and the task" (p.321).

Closer examination of individual Trial Block data however indicated that several participants did indeed alter strategies following repeated exposure to the surface dynamics. Participants may have been exploring the utility of different control strategies in response to the altered support surface properties. That is, participants may have explored the individual-support surface dynamics in an attempt to identify the most appropriate or stable coordination pattern. Exploratory behavior in response to variations in support surface properties has been previously reported (Horak & Nashner, 1986; Stoffregen et al., 1997). Horak and Nashner (1986) using a perturbation paradigm demonstrated that young adult participants utilized complex combinations of ankle and hip strategies when initially exposed to alterations in support surface properties. Following repeated exposure to surface dynamics the complex exploratory strategies were progressively changed to surface-specific ankle or hip control strategies.

Head Stabilization

Study results failed to support the hypothesis that head stability would be maximized on the compliant and firm, low-friction surfaces. In contrast, the results indicated that both groups of older adult participants demonstrated the least stable head trajectory when leaning forward through their stability region during Trial Block I (i.e., during initially exposure) on the compliant support surface. These findings suggest that when somatosensory inputs were altered due to changes in the support surface dynamics (i.e., support surface with low resistance to ankle torques), head stability in older adults was compromised (Alexander et al., 1992). Repeated exposure to compliant support surface dynamics, however, resulted in an increase in head stability that more closely approximated head trajectory profiles evidenced on the firm, high- and low-friction surfaces. This finding suggests that participants identified and adapted the control of head motion to the compliant support surface constraints.

The increase in head stability on the compliant surface from Trial Block I to Trial Block II occurred, however, despite any observable changes in postural control strategies. Since support surface dynamics constrained postural coordination, participants needed to utilize alternative strategies to stabilize head motions. Although the present investigation did not examine compensatory motions associated with head control, one possible mechanism by which participants may have controlled motions of the head was to adopt a head-stabilization-in-space strategy. Recent findings by DiFabio and Emasithi (1997) indicated that the head-stabilization-in-space strategy was associated with increased head control during the performance of difficult postural tasks. Further examination into the relationship of head-stabilization-in-space strategy and the stability of head motions across variations in support surface properties is warranted.

Previously reported findings have indicated that both young and older adults demonstrate increased head stability when postural control is maintained on unstable and/or unpredictable support surfaces. The data from the present experiment however, do not provide support for these findings (DiFabio & Emasithi, 1997; Lekhel et al., 1994; Pozzo et al., 1992). Although, the RMS head acceleration values associated with task performance on the firm, low-friction surface did indicate greater head stability when compared with head motions evidenced on the firm, high-friction surface, the observed

differences failed to reach a level of statistical significance. The lack of statistical differences in RMS head acceleration values for these two surface conditions may be attributed to similarities in the perceptual inputs (e.g., visual, vestibular and somatosensory) available from the individual-surface dynamics. That is, functionally the frictional and torque supporting properties of the two support surfaces may have been similar thereby providing participants with similar perceptual information during task performance. As suggested by DiFabio and Emasithi (1997), these perceptual inputs serve a critical role in the control of head stability during volitional movements.

RMS head acceleration data from the present investigation also failed to support the hypothesis that the degree of head stability would be greater for the older adults at-high risk for falls across all support surface conditions. A closer examination of the data, however, indicated that the degree of head stability during task performance indeed tended to be greater for the group of older adults at-high risk for falls across each of the three test surfaces. However, the low number of participants representing each group may have influenced these findings. Specifically, the calculated statistical power to detect differences between groups (i.e., level of postural stability) was .085. Thus, an increase in statistical power through larger sample sizes may strengthen the observed group differences. DiFabio and Emasithi (1997) in a previous investigation have demonstrated that the magnitude of head stability, when performing difficult postural tasks, was significantly greater in a group of older adults (with no reported vestibular symptoms) when compared with a young adult sample.

Conclusions

Data from the present investigation indicate that postural control strategies and the degree of head stability exhibited by older adults when leaning forward through the stability region are influenced by the dynamic properties of the support surface. The results suggest that, irrespective of level of postural stability, older adults are able to adapt the coordination and control of body segments to variations in support surface constraints. Although variations in support surface properties constrained the coordinated motions and control of body segments for the majority of older adult participants in the study, several participants demonstrated robust control across the three

different support surface conditions. Similarities in the observed postural control strategies across variations in surface dynamics may be attributed to the different support surface conditions being able to support similar coordination patterns and/or study participants "relaxing" task goals to maintain postural stability during task performance.

Variations in the support surface properties also influenced the degree of head stabilization during the task performance. Support surface dynamics in which somatosensory inputs were altered (e.g., compliant surface conditions) resulted in compromised head stability irrespective of the older adults' level of postural control. The degree of head stabilization increased, however, after repeated exposure to the compliant surface dynamics. Since postural control strategies did not change across trial blocks, the increase in head stability may be attributed to compensatory strategies associated with the control of head motions (i.e., adopted a head-stabilization-in-space strategy) (DiFabio & Emasithi, 1997). These findings indicate that further examination of head control strategies across variations in support surface dynamics may be fruitful.

Adaptations in the coordination and control of posture in response to support surface constraints provide support for the theory of perception and the control of bodily orientation (Riccio & Stoffregen, 1988). Specifically, the present findings indicate that the coordinated movements of the body and its segments emerge from the constraints imposed by the interaction of the support surface, the task, and the individual. In conclusion, to further our understanding of postural control as it relates to falls and fall risk among older adults, researchers and practitioners should study movements and actions as individuals interact with various support surface properties while performing goal-directed behavior(s).

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Chapter 3

SUPPORT SURFACE CONSTRAINTS ON THE COORDINATION AND CONTROL OF GAIT IN OLDER ADULTS

Sean Clark

Abstract

Although age-associated changes in walking patterns have been well documented, these studies have been conducted, almost exclusively, on a firm, level, slip-resistant surface. Gait evaluations performed on a single test surface do not address the more ecologically relevant issue of whether older adults can alter their walking patterns to support surface conditions similar to those encountered during everyday tasks. Within the theoretical framework of the perception and control of bodily orientation (Riccio & Stoffregen, 1988), the ability to accommodate variations in surface properties and maintain a functional level of postural stability requires that an individual perceives the constraints of the surface for action and adapts the control and coordination of body motion to these identified constraints. The purpose of this study was to examine the coordination and control of actions while 8 healthy older adults and 8 older adults identified at-high risk for falls walked at preferred speeds across each of 3 surfaces with different support properties: rigid, high friction; compliant; and rigid, low friction. Body kinematics for trials 1-3 (T1) and 10-12 (T2) from 12 completed trials on each surface were analyzed using three-dimensional (3-D) video analysis. Measures of the control and coordination of lower extremity movements as well as the degree of head stability in the sagittal plane were examined for differences across level of postural stability, surface conditions, and repeated exposure to surface dynamics. Results indicated that variations in support surface properties were associated with observed changes in the coordination and control of lower extremity movement patterns, as well as the degree of head stability in the sagittal plane during the gait cycle. Additionally, significant differences in the degree of head stability and control of lower extremity joints were evidenced when comparing healthy older adults with the at-high risk for falls group. In conclusion, findings from the present investigation provided empirical support for the theory of perception and control of bodily orientation by demonstrating that gait patterns adopted by older adults while walking were indeed influenced by both variations in support surface properties and level of postural stability.

Introduction

Many community-dwelling older adults, as well as elderly residents in long-term care facilities, experience difficulties performing tasks or activities that involve walking. These difficulties often emerge due to age- and/or disease-related changes in the perceptual (e.g., visual, vestibular, somatosensory) and action (e.g., musculoskeletal) systems (Gibbs, Hughes, Dunlop, et al., 1993; Sudarsky & Ronthal, 1992). Such changes may compromise an older adult's ability to appropriately perceive relevant information specifying the properties of the environment and/or adapt the control of movement to identified environmental hazards (Patla, Frank & Winter, 1990). Variations in support surface properties (e.g., different degrees of surface compliance or coefficients of friction), therefore, may have negative implications for the coordination and control of bodily orientation during goal-directed behavior (Tideiksaar, 1990). Thus, falls or the increased risk for falling in older adults while walking have often been attributed to a diminished ability to either anticipate or adapt to the demands and challenges in the environment (Cali & Kiel, 1995; Mayo, Korner-Bitensky & Levy, 1993; Prudham & Evans, 1981, Tinetti, Speechley & Ginter, 1988; Tinetti, Doucette, & Claus, 1995).

During goal-directed behavior, such as walking, the effective control and coordination of movements for action is influenced or constrained by the physical dynamics of the environment (e.g., friction, firmness, extent, and/or inclination of the support surface) (Horak & Nashner, 1986; Stoffregen & Flynn, 1994). As such, the coordinated movement of body segments effective for walking on one surface may not be appropriate for a surface with different support properties. The ability to identify surface constraints and adapt the control of movement accordingly is necessary to maintain postural stability and minimize the risk for falls (Patla et al., 1990; Riccio & Stoffregen, 1991). From a theoretical framework of perception and control of bodily orientation, Riccio and Stoffregen (1988, 1991) suggest that surface constraints on action are identified or perceived through the kinematics of an individual's interaction with the environment (i.e., individual-environment dynamics). That is, as a person walks across a support surface, changes in body kinematics (e.g., head and limb motions) provide perceptual information to the visual, vestibular, and somatosensory systems necessary for

identifying the layout of the environment, traversability of the support surface, and utility of different movement strategies for the control of action.

Changes in the sensitivity of the perceptual systems to stimulation (e.g., peripheral neuropathies or visual-vestibular impairments) and/or stimulation of the perceptual systems via sources independent of the surface dynamics (e.g., postural or intentional tremor) alter the relationship or pattern of perceptual information that results from the kinematics of the individual-environment interaction (Pozzo, Berthoz, Lefort & Vitte, 1991; Riccio & Stoffregen, 1991). Failure to detect the new relationship between the patterns of perceptual stimulatión and the individual-environment dynamics may result in an inability to correctly perceive the support surface dynamics, including appropriate postural control strategies. Since the effective control of posture is constrained by support surface dynamics, persons who experience age-related changes in the sensitivity of the perceptual systems are often at an increased risk for falls (Prudham & Evans, 1981; Tinetti et al., 1983).

Although the relationship between age-associated changes in the sensitivity of the perceptual systems (e.g., impairments) and fall-risk have been well documented (Gibbs, Hughes, Dunlop, et al., 1993; Sudarsky & Ronthal, 1992, Tinetti et al., 1983), the relationship between the relative stability of the perceptual systems and postural instability has received much less empirical investigation. Recent findings, however, have indicated that head stability is essential to maintaining the integrity of visual and vestibular inputs (Assaiante & Amblard, 1990; Pozzo, Berthoz & Lefort, 1990; Pozzo, Levik & Berthoz, 1992). For example, in complex environments and/or environments in which perceptual inputs were not available (e.g., dark room), younger adults exhibited an increase in head stability when walking (Assaiante & Amblard, 1990). Such increases in head stabilization when maneuvering in complex environments may be critical to identifying the dynamics of the individual-environment interaction. That is, self or internally generated motions of the head may give rise to stimulation of the visual and vestibular systems that preclude detection of external perturbations and/or the dynamic properties of the environment (Holt, Jeng, Ratcliffe & Hamill, 1995; Horak & Macpherson, 1995, Pozzo, et al., 1991). Consequently, variability in upper-body and head accelerations while walking may negatively influence the utilization of visual and

vestibular information necessary for the control of bodily orientation (Riccio & Stoffregen, 1991). For older adults, especially individuals at-high risk for falls, control of head and trunk motion may serve to optimize the integrity of the visual and vestibular inputs and thereby facilitate perception of the individual-environment dynamics (DiFabio & Emasithi, 1997). Such minimization of head instability may be crucial in reducing further risk for falls in older adults identified with compromised postural stability.

Although perception of environmental constraints are critical to postural control, the accomplishment of task goals also requires that persons coordinate movements or actions according to the constraints imposed by both the specific task and environment (Riley, Mitra, Stoffregen & Turvey, 1997; Riccio & Stoffregen, 1991; Marin, Bardy, Baumberger, Fluckiger & Stoffregen, 1997). That is, given task and the environmental constraints, individuals must identify the appropriate means by which to control and coordinate the multiple or redundant degrees of freedom (e.g., joints) associated with the necessary actions for the accomplishment of task goals (Riccio & Stoffregen, 1988; Vereijken, vanEmmerik, Whiting & Newell, 1992). Goal-oriented tasks that require unfamiliar and/or complex movements are often initially controlled by reducing the number of degrees of freedom available at an anatomical joint (Vereijken, et al., 1992). Such increased rigidity at the joint provides a control strategy that enables a person to perform the task by minimizing demands on the control of movement. As individuals become more skilled at accomplishing task goals, the degree of control at the joint is often reduced (i.e., releasing degrees of freedom) (Vereijken, et al., 1992).

Although age-associated changes in walking patterns have been well documented (Craik, 1989; Eble, Thomas, Higgins & Colliver, 1991; Ferrandez, Pailhous & Durup, 1990; Finley, Cody & Finizie, 1969; Murray, Kory & Clarkson, 1969; Oberg, Karsznia & Oberg, 1993; Winter, Patla, Frank & Walt, 1990; Yack & Berger, 1993), these studies have been conducted, almost exclusively, on firm, level, slip resistant surfaces. Activities performed during daily living, however, often require that community-dwelling older adults interact and maintain dynamic balance across a wide variety of surfaces with different support properties (e.g., friction, firmness, extent and inclination) (Cutson, 1994; Tideiksaar, 1990). Consequently, gait evaluations performed on a single test surface do not address the more ecologically relevant issue of whether older adults can

alter their walking patterns to support surface conditions similar to those encountered during everyday tasks (Stoffregen & Flynn, 1994). Since falls often occur as a result of the inability to perceive and/or alter movement patterns to variations in support surface properties (e.g., slippery surface), assessments of walking patterns across multiple surfaces (i.e., different support properties) would provide insight into additional factors associated with increased fall-risk among elderly persons.

The purpose of this second study was to examine the coordination and control of action while both healthy older adults and older adults identified with compromised postural stability walked across a variety of surfaces with different support properties. Specifically, the study had four primary objectives: (A) determine whether the degree of head stability is different across variations in support surface dynamics, (B) contrast the degree of head stability exhibited by healthy older adults with that exhibited by older adults who were identified at-high risk for falls as they walked across three different support surfaces, (C) determine whether coordination and/or control of lower extremity movement patterns differ across different support surfaces, and (D) contrast the adaptive coordination and control of lower extremity movement patterns for healthy older adults to that of older adults at-high risk for falls following initial and repeated exposure to the support surface dynamics. As an extension of these specific objectives it was further hypothesized that: (A) head stability would be maximized when persons walked on compliant and low friction surfaces, (B) the degree of head stability evident in persons at-high risk for falls would be significantly greater than that observed for healthy older adults across all test surfaces, (C) variations in support surface properties would result in observable differences in the coordination and control of lower extremity movement patterns, and (D) healthy older adults, in contrast to older adults at-high risk for falls, would demonstrate significantly better adaptive coordination and control of movements across variations in support surface dynamics.

Methods

Participants

Eight healthy older adults and eight older adults identified at-high risk for falls (76-86 years), were recruited from the community to participate in the present study.

These persons were identified from a pre-study screening that provided a comprehensive clinical assessment of balance and mobility. Assessments in the pre-study screening included a: (a) background and medical history questionnaire (see Appendix A), (b) Berg Balance Scale (Berg, 1993) (see Appendix B), (c) Limits of Stability Test® (LOS) conducted at 100% of the participant's theoretical limits of stability, and (d) Sensory Organization Test®(SOT). Prior to participation, all persons provided written informed consent in accordance with policies outlined by the University Institutional Review Board at Oregon State University (see Appendix C).

Of 55 older adults who completed the pre-study screening tests, eight persons met the specific eligibility criteria for the healthy group and nine individuals met criteria for the group identified at-high risk for falls (one person withdrew prior to the start of the study due to an unexpected illness). Eligibility criteria for the healthy older adult group were as follows: (a) living independently in the community (non-institutionalized setting), (b) no persistent problems with dizziness or postural unsteadiness, (c) no musculoskeletal impairments, (d) a performance score greater than 50/56 on the Berg Balance Scale, (e) a mean Maximum Excursion value on the Limits of Stability Test greater than 80% LOS, and (f) an average equilibrium score on the Sensory Organization Test greater than 70. In contrast, criteria for the group identified as being at-high risk for falls were as follows: (a) living independently in the community (non-institutionalized setting), (b) a self-report of postural instability and/or repeated falls in the past 2 years, (c) performance scores equal to or below 50/56 on the Berg Balance Scale, (d) a mean Maximum Excursion value on the Limits of Stability Test less than 80% LOS, and (f) an average equilibrium score on the Sensory Organization Test equal to or less than 70. Persons with diseases known to adversely affect balance and mobility (e.g., Parkinson's disease, multiple sclerosis, CVA, diabetes), permanent orthopedic impairments (e.g., fused joints, amputation) and/or visual deficits not correctable with lenses were excluded from participation in the study.

The criteria scores used to establish group eligibility were determined from pilot study data. Pilot data from the performance scores of 144 community-dwelling older adults (with and without a self-reported history of falls) between the ages of 65 and 92

years were used to calculate means and standard deviations for the SOT and LOS tests. The calculated mean equilibrium score of 70 and mean Maximum Excursion score (i.e., composite score from the 8 targets) of 80% LOS, from these pilot data, were used as the eligibility cut-off scores for the SOT and LOS tests, respectively. In addition, the criteria score of 50/56 for the Berg Balance Scale was selected based on previously reported performance scores from a group of older adults at-risk for falls (Rose & Clark, 1995).

Table 2.1. Group eligibility criteria data.

	Older adults at-high risk for falls	Healthy older adults
Number (N)	8	8
Male	3	4
Female	5	4
Mean Age (yrs.)	80.9	80.6
Age Range (yrs.)	76-86	77-83
Berg Score	48.4 (1.92) *	54.0 (1.85)
Maximum Excursion (% LOS)	69.6 (9.45) *	87.5 (4.25)
Equilibrium Score (% stability)	63.9 (6.36) *	79.5 (4.81)

Performance scores for older adults at-high risk for falls compared to healthy older adults group (* significant at $p \leq .001$ level).

The participants in the present investigation were concurrently involved in a study examining the influence of support surface properties on the control and coordination of posture during upright stance. To avoid possible confounds related to support surface experience, participation in the two studies was counterbalanced for the walking and standing tasks and the presentation order of surfaces within each study was randomized. Means and standard deviations for the group criteria data for the two groups of older adults are presented in Table 2.1. Separate one-way ANOVAs were conducted for each criteria measure, as well as age, to ensure that groups did not differ in age but were

significantly different with respect to each of the three performance measures ($p \leq .001$).

Support surfaces

Lower extremity joint coordination and head stability were evaluated while each participant walked at a preferred-speed on each of three different support surfaces: (A) firm, high-friction surface, (rubber matting), (B) compliant surface, (open cell, polyurethane foam) and, (C) firm, low-friction surface, (vinyl flooring coated with a thin film of glycerin-water solution).

Preparation

A standard link-segment model that included the foot, shank, thigh, trunk, and head was developed to assist in the analyses of upper- and lower-body kinematics while participants walked across each of the three support surfaces. Retroreflective, spherical markers (15mm in diameter) were secured with double-sided adhesive tape to bony landmarks, on the right side of the body, corresponding to the fifth metatarsal (toe), lateral portion of the calcaneous (heel), lateral malleolus (ankle), lateral femoral epicondyle (knee), greater trochanter (hip), greater tubercle of the humerus (shoulder), and temporal process of the mandible (head). Participants were provided with lycra bike shorts and a tank top to wear during all testing procedures.

The three-dimensional (3-D) motions of the reflective markers during each test trial were recorded using two synchronized video cameras (Panasonic DT5100) and external video tape recorders. The cameras were configured with exposure time of 0.001 seconds and frame rate of 60Hz.. The video cameras were positioned 5.7 and 5.4 meters from the middle of a 6 meter walkway with an angle of 95° between the cameras' optical axes (see Figure 2.1 for experimental setup). The camera placement provided an unobstructed right sagittal view of each participant as she/he walked across each surface. Prior to data collections, a calibration structure was positioned in the middle of the walkway and video recordings were obtained. Digitized data of the calibration structure were used in subsequent calculations to calibrate the measurement volume.

Experimental Procedures

Following marker preparation procedures, the participant was secured to an overhead harness system. The harness system was designed to allow the participant to walk freely (i.e., unrestricted) yet prevent the occurrence of any falls. The harness system consisted of a vest-harness attached at the shoulders via carabiners and kevlar webbing to a moveable trolley (four steel rollers with ball bearings) that slid within two aluminum sliding door tracks (30' long). The overhead aluminum tracks were permanently fixed to a wooden ceiling-beam in the research laboratory.

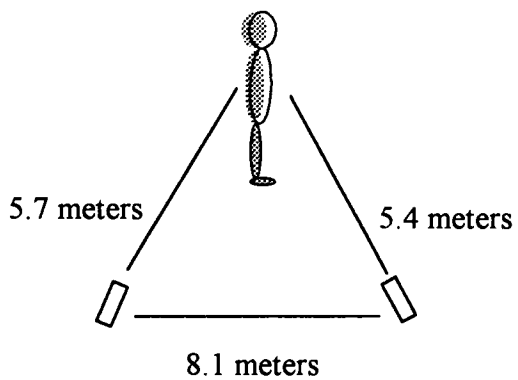

Figure 2.1. Experimental set-up.

Assistance was provided to the participant while she/he stepped onto the test surface. Once the participant was able to stand independently on the surface, she/he was instructed to maintain a stable upright posture while walking barefoot across the surface at a comfortable speed. As the participant walked across the 6 meter walkway an assistant followed slightly behind and to the left side of the participant (i.e., off the support surface) to provide assistance should support have been required and/or requested by the participant. Upon completion of a test trial, the participant was assisted off the test surface and she/he returned to the start position. A 30-second rest period was provided prior to the start of the next trial. Throughout the testing a chair was placed next to the test surface to allow the participant an opportunity to sit down during each rest period.

For each test surface, participants completed 12 trials and these data were used for subsequent analyses. A three-minute rest period was provided prior to the start of test trials on a new support surface. Presentation of test surface conditions was randomized for each participant.

Data Reduction

From the recorded video data of the 12 test trials on each of the three surfaces, the video image of each camera view for trials 1 through 3 (trial block I) and trials 10 through 12 (trial block II) was autodigitized at 60 Hz using the Peak Performance Motion Analysis System (software version 5.3). Analyses were limited to trial block I (n=3) and trial block II (n=3) in order to obtain trials that represented the participant's initial response to the support surface and the participant's response after repeated exposure to the surface properties, respectively. High frequency noise in the raw digitized coordinate data for each camera was filtered using a fourth-order, zero-phase Butterworth low-pass filter prior to calculations of the 3-D coordinates. The appropriate cutoff frequency for each data point in the three dimensions was determined using the Peak Performance Optimal Filter option, which is based on the Jackson Knee Method (Jackson, 1979). Three-dimensional position-time coordinates from the two camera views were determined using the filtered coordinate data and the Direct Linear Transformation (DLT) technique (Abdel-Aziz & Karara, 1971). The 3-D coordinate calculations were used to derive kinematic parameters (i.e., positions, velocities, and accelerations) utilized in the subsequent analyses of the coordination and control of upright posture while walking.

Measures of Interest

Kinematic parameters were calculated from the video data for each participant in order to characterize lower extremity movement patterns during a single stride identified in the middle portion of the walkway. Calculations of the sagittal plane angular positions for the ankle (relative angle between the foot segment and the shank), knee (relative angle between the shank and the thigh), and hip (relative angle between the thigh and the trunk) were derived from the 3-D filtered data. Standard deviations of each joint angular

position time-series data were derived to provide an indication of the degree of control at each joint. The calculated variability in the joint angular position data (i.e., ankle, knee, and hip) for each stride cycle was used to assess changes in the control of individual degrees of freedom (Vereijken, et al., 1992). Low standard deviation values were indicative of high levels of control at the respective joint (i.e., the joint being rigid or fixed during the stride cycle). In contrast, high standard deviation values were indicative of a release of rigid control. Variability in the angular position data for each joint was determined as the mean standard deviation value from the three stride cycles corresponding to trial block I and II across each of the three surface conditions.

Additionally, angle-angle diagrams of the hip and knee angular position data were constructed to provide a qualitative assessment of the coordination of lower extremity movement patterns while persons walked across each of the three test surfaces (Enoka, Miller, & Burgess, 1982; Winstein & Garfinkel, 1989). Prior to constructing the angle-angle diagrams, joint angular position time-series data were normalized from 0-100% for one complete stride. Linear interpolations were used in the normalization procedure to estimate data points for the common time intervals. Normalized joint angular position data (i.e., hip and knee) were ensemble-averaged for the three trials within both trial block I and trial block II. A single ensemble-averaged angle-angle diagram was constructed to provide a representative graphic display of the relative coordination patterns of the lower extremities (i.e., hip and knee joints) during the gait cycle for the respective trial blocks on each test surface. This procedure was performed for each participant across the three test surfaces, resulting in a total of six angle-angle diagrams for each participant. Qualitative assessments of the angle-angle diagram characteristics; including shape and conjoint range of motion were performed for each angle-angle diagram. Conjoint range of motion refers to the total change in angular position in the lower extremity as a function of variation in both the hip and knee joint angular positions during the gait cycle. Qualitative assessments provided an indication of the characteristics of intralimb coordination for each participant across each test surface (Hershler & Milner, 1980; Winstein & Garfinkel, 1989).

The angle of the foot at foot-surface contact (relative angle between the plantar surface of the foot segment and the support surface) was also calculated to assist in

identifying adaptations in the control of movement while participants walked across the various test surfaces. The foot angle relative to the support surface was used to provide an indication of the shear and normal forces associated with foot impact (i.e., foot-surface contact). Foot contact was derived through visual inspection of position-time graphs depicting the linear position of the heel marker in the vertical and anterior-posterior (A/P) planes across time (i.e., stride cycle). The time value corresponding to changes in linear position of less than 1 centimeter in both the vertical and A/P planes was used as the temporal measure to identify foot contact. Prior to statistical analyses, measures of foot angle were averaged for both trial block I and trial block II for each test surface.

In addition to the calculations of lower extremity kinematics, Root Mean Square (RMS) acceleration values were derived from the calculated linear acceleration profiles of the head marker in the sagittal plane (i.e., anterior-posterior direction) for each stride cycle. The RMS head acceleration values were used to provide a measure of the degree of stability or consistency of motion associated with movements of the head segment. For statistical analyses, measures of RMS head acceleration were averaged for the three stride cycles from both trial block I and trial block II for each of the three test surfaces. Consequently, averaged RMS head acceleration values were used to indicate the degree of head stability while persons walked across each support surface (Prince, Winter, Stergiou, & Walt, 1994).

Statistical Design & Data Analyses

A 2 x 3 x 2 (Group x Surface Condition x Trial Block) between-within design with repeated measures on the last two factors was implemented to address three study objectives; A) whether head stability and/or the coordination and control of the lower extremities differed as a function of support surface conditions, B) contrast the degree of head stability between the high-risk for falls group with the healthy older adult group across all surface conditions, and C) contrast the adaptive coordination and control of lower extremity movement patterns exhibited by healthy older adults with those exhibited by the high-risk for falls older adults following initial and repeated exposure to the support surface dynamics. Two separate univariate 2 x 3 x 2 (Group x Surface Condition x Trial Block) repeated measures analyses of variance (RM ANOVA) were

used to determine how the degree of head consistency and coordination of lower extremities (i.e., the relative angle of foot at heel strike) were influenced by the level of postural stability (i.e., healthy older adults versus older adults at-high risk for falls), support surface condition, and repeated exposure to support surface properties. Statistical significance was determined using an alpha level set at $p \leq .05$. Additionally, significance in the surface main effect and/or higher order interactions were further assessed using Scheffe's post-hoc comparisons.

A 2 x 3 x 3 x 2 (Group x Joint Angle x Surface Condition x Trial Block) repeated measures analysis of variance (RM ANOVA) was also performed on the hip, knee and ankle joint angular position time-series standard deviations. This analysis was conducted to determine differences in the degree of control (i.e., variability) of lower extremity joint activity during a stride cycle for the main effects of group, joint angle, surface condition, and trial block as well as all higher order interactions. Significance in any of these main effects and/or higher order interactions was further assessed using Scheffe's post-hoc comparisons.

Results

Head Stability

A 2 x 3 x 2 (Group x Surface Condition x Trial Block) univariate RM ANOVA was conducted using RMS head acceleration data to determine whether the degree of head stability was different across groups (study objective B), support surface condition (study objective A), and repeated exposure to support surface properties. The RM ANOVA output table and the corresponding power estimates and eta squared values are presented in Appendix F. Results of the RM ANOVA yielded significant main effects for the Group ($F(1,14) = 5.025$, $p \leq .05$) and Surface Condition ($F(2,28)=5.45$, $p \leq .02$) factors only (see Figure 2.2). A subsequent comparison of group means indicated that head trajectory motions were significantly more stable (i.e., lower RMS head acceleration values) for the group of older adults identified at-high risk for falls when contrasted with the healthy older adult group across all support surface conditions. In addition, Scheffe' post-hoc comparisons conducted for the significant Surface Condition main effect

indicated that the RMS head acceleration values derived from the compliant surface condition were significantly larger than those for the firm, low-friction surface. No differences in RMS head accelerations were determined between the firm, high-friction surface and the compliant, or firm, low-friction surfaces.

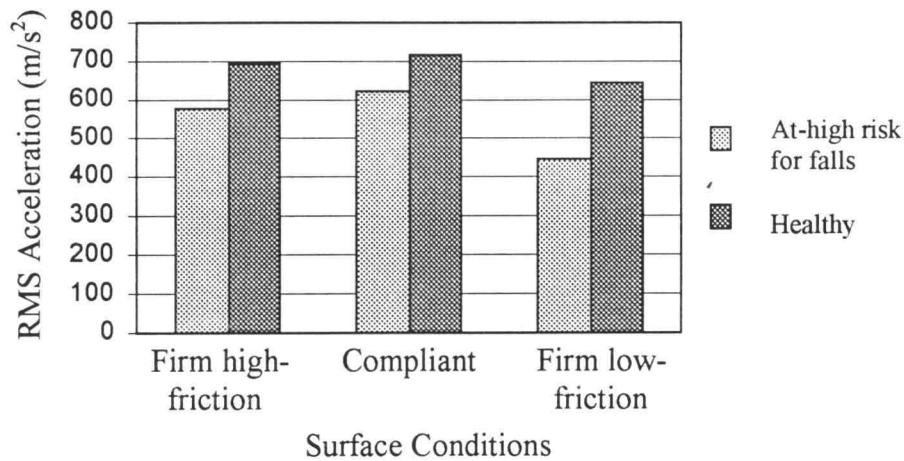

Figure 2.2. Averaged RMS head acceleration values for at-high risk for falls and healthy older adult groups across surface conditions.

Control of lower extremity movement patterns

A 2 x 3 x 3 x 2 (Group x Joint Angle x Surface Condition x Trial Block) RM ANOVA using standard deviations of the joint angular position time-series data was used to determine whether the control of lower extremity movement patterns were different across support surface conditions (study objective C)(see Table 2.2). Results of this analysis yielded significant main effects for Joint Angle ($F(2,28) = 573.87$, $p \leq .001$), and Surface Condition ($F(2,28) = 390.70$, $p \leq .001$). These main effects, however, were superseded by a significant interaction effect between Joint Angle and Surface Condition ($F(4,56) = 156.24$, $p \leq .001$) (see Figure 2.3). Results of the follow-up Scheffe's comparison indicated that the difference in the variability between the knee joint angular motion and that of both the hip and ankle joints on the firm, low-friction surface was

significantly smaller than the observed differences in the variability between the knee joint and the hip and ankle joints on both the compliant and firm, high-friction surfaces.

The 2 x 3 x 3 x 2 (Group x Joint Angle x Surface Condition x Trial Block) RM ANOVA also addressed study objective D, which was to contrast the degree of lower extremity joint control demonstrated by the healthy older adults with that of older adults at-high risk for falls across both variations in support surface conditions and trial blocks. Results of the RM ANOVA yielded a significant Group main effect ($F(1,14) = 15.08$, $p \leq .002$). This finding indicated that the variability associated with joint angular motions during the gait cycle for the healthy older adults was significantly larger than that for older adults at-high risk for falls across both joint angles (i.e., ankle, knee and hip) and support surface conditions. The RM ANOVA output table containing both main and interaction effects with corresponding power estimates and eta squared values are presented in Appendix G.

Table 2.2. Means of standard deviations of the joint angular position time-series data for at-high risk for falls and healthy older adults across surface conditions.

	Surface Conditions		
Joint Angle	Firm, high-friction	Compliant	Firm, low-friction
Hip			
At-high risk	5.85	11.79	4.44
Healthy	6.40	13.18	6.32
Knee			
At-high risk	16.63	23.41	8.97
Healthy	18.46	25.52	10.92
Ankle			
At-high risk	6.51	6.93	4.43
Healthy	8.33	9.39	6.04

Means of standard deviation values derived from the joint angular position time-series data for each joint for each surface condition, collapsed across trial blocks.

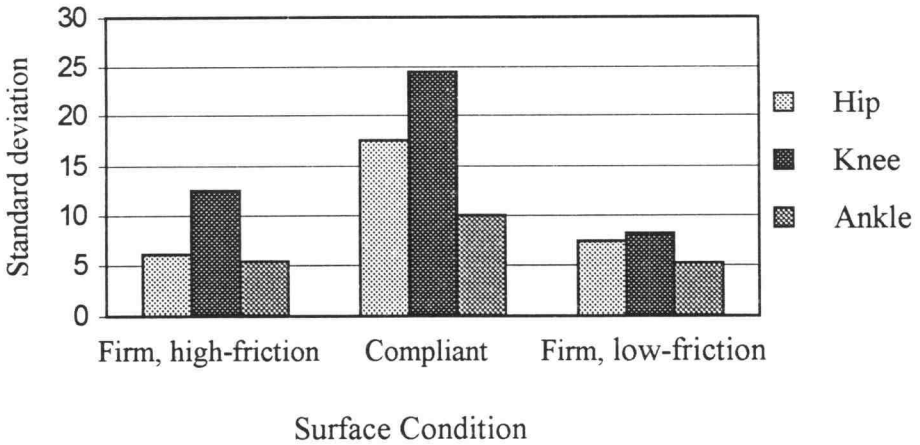

Figure 2.3. Mean joint angle standard deviations for the hip, knee and ankle joints across surface conditions collapsed across groups and trial blocks.

Coordination of lower extremity movement patterns

The influence of level of postural stability, support surface condition, and repeated exposure to support surface properties on the coordination of lower extremity movement patterns was also addressed qualitatively by constructing hip-knee angle-angle diagrams for each participant as a function of Surface Condition and Trial Block. Visual inspection of each of the 96 hip-knee angle-angle diagrams revealed notable differences in lower extremity kinematic patterns when older adults walked across surfaces with different support properties (see Figures 2.4 - 2.6). These representative figures illustrate the large conjoint range of motion at the hip and knee joints when persons walked on the compliant support surface. In contrast to performance on the compliant surface, reductions in hip-knee conjoint range of motion was apparent when participants walked across the firm, low-friction surface.

Qualitatively, the patterns (i.e., shapes) of the angle-angle diagrams within each surface condition were similar irrespective of the participants' level of postural stability. However, the healthy older adult group generally demonstrated a larger conjoint range of motion across each of the three different support surfaces. Following repeated exposure to the dynamics of the firm, low-friction surface, several of the study participants demonstrated increased conjoint range of motion in the hip and knee joints during the gait

cycle. It should be noted however, that this observation was not apparent across all study participants and the increases were equally distributed between the two test groups.

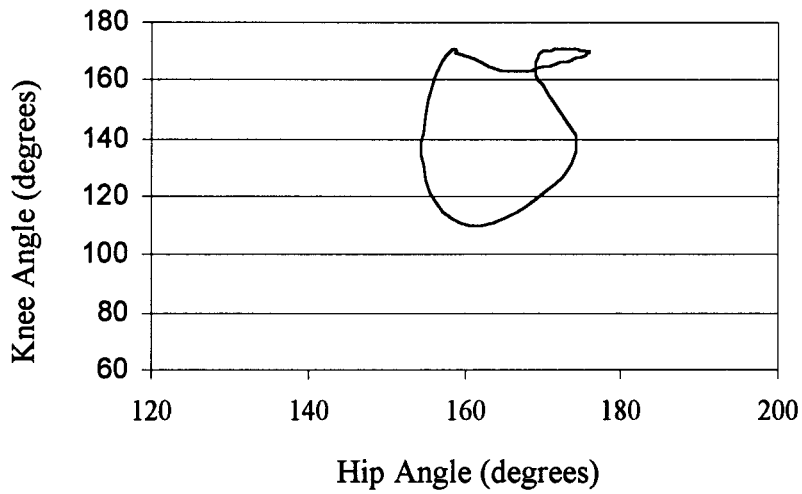

Figure 2.4. Representative hip-knee angle-angle diagram for the firm, high-friction surface.

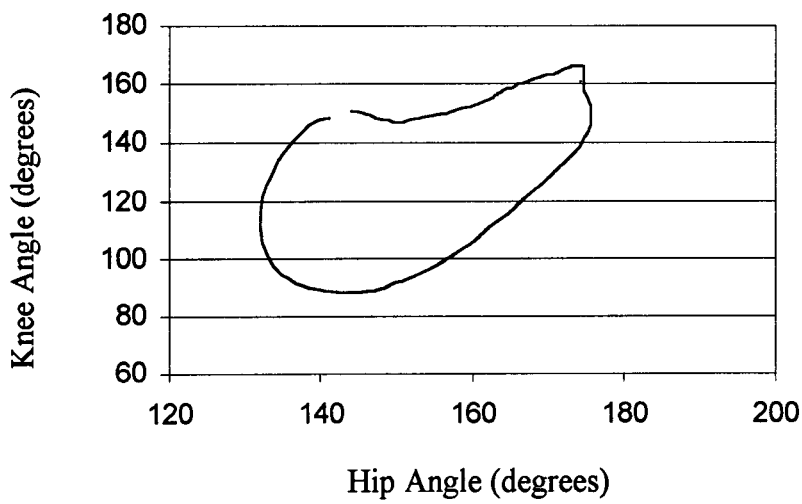

Figure 2.5. Representative hip-knee angle-angle diagram for the compliant surface.

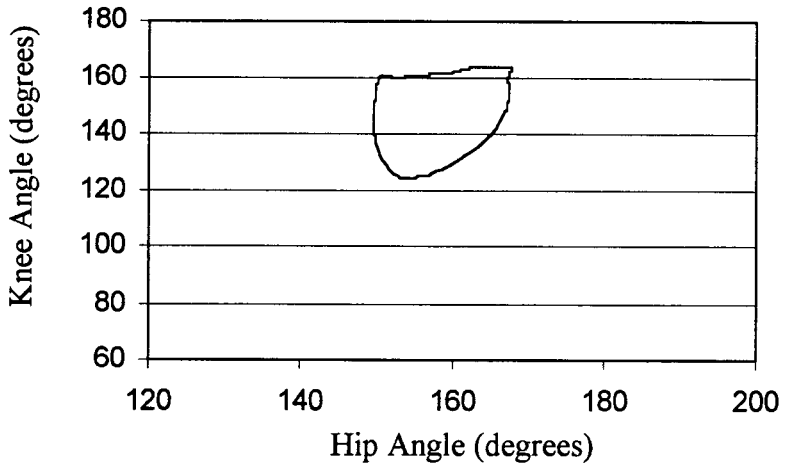

Figure 2.6. Representative hip-knee angle-angle diagram for the firm, low-friction surface.

To further determine whether the coordination of lower-extremity movement patterns were different across support surface conditions (study objective A) and/or if group differences existed following repeated exposure to support surfaces dynamics (study objective C), a 2 x 3 x 2 (Groups x Surface Condition x Trial Block) RM ANOVA was conducted using the dependent variable mean relative foot angle at contact with surface. Results of this analysis indicated a significant main effect for Surface Condition ($F(2,28) = 31.10$, $p \leq .001$) and a significant two-way interaction between Surface Condition and Trial Block ($F(2,28) = 4.59$, $p \leq .02$). The Surface Condition x Trial Block interaction was further examined using a Scheffe's post-hoc test of multiple comparison. Results indicated the interaction effect was attributed to a significant decrease in the difference between the foot angle at surface contact on the firm, high-friction surface and the firm, low-friction surface across Trial Blocks (see Figure 2.7). The RM ANOVA output table with corresponding power estimates and eta squared values are presented in Appendix H.

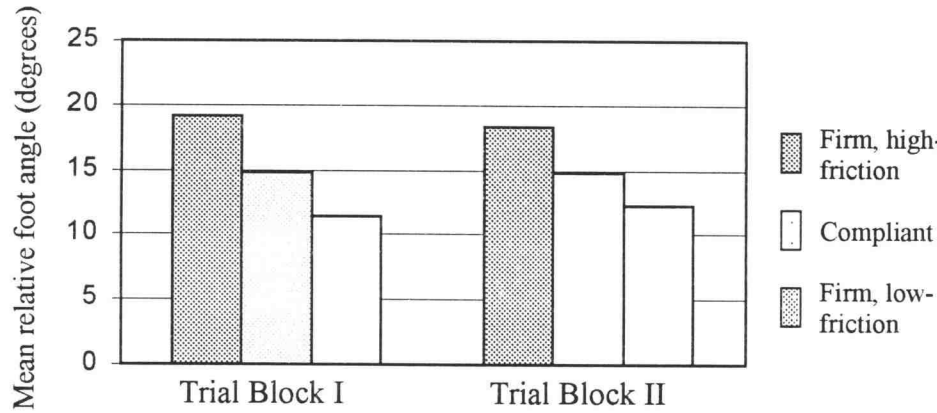

Figure 2.7. Mean relative foot angle at surface contact for each surface condition across trial blocks collapsed across groups.

Discussion

Head Stabilization

Findings from the present investigation failed to provide support for the proposed hypothesis that head stability would be maximized when persons walked on both the firm, low-friction and compliant surfaces. In fact, motions of the head trajectory in the sagittal plane were least stable for test trials conducted on the compliant support surface. These results may indicate that the older adult participants experienced difficulties in maintaining head stability due to perturbations associated with the "uneven" or compressive properties of the compliant surface. Stoffregen and Flynn (1994) have recently suggested that when individuals walk on a compliant or deformable support surface the compressive forces associated with foot contact "... should be dissipated more slowly, producing a temporally extended acceleration of low magnitude" (p.60). These authors further suggest that the rate of acceleration associated with the compressive forces may change following initial contact with the support surface (Stoffregen & Flynn, 1994). Since the body is a multi-linked segment, these changes in accelerations (i.e., perturbations) will be transferred to the head unless attenuated in the segmental kinetics and/or kinematics (Holt et al., 1995; Prince et al., 1994; Stoffregen & Flynn, 1994). Consequently, the observed difficulties in maintaining head stability when walking on the compliant surface may be attributed to the inability of the older adult

participants to: a) appropriately perceive the support surface dynamics for the control of action (Riccio & Stoffregen, 1991; Stoffregen & Flynn, 1994), b) increase the relative phasing of lower extremity coordination patterns to minimize perturbations associated with foot strike and toe-off (Holt et al., 1995; Jeng, Holt, Fetters & Certo, 1996; Kasser & Clark, 1997) and/or c) progressively attenuate accelerations through the upper-body (Prince et al., 1994). Additional study of compliant surface constraints on head stability among older adults is required to more clearly elucidate the mechanism(s) contributing to head instability. Moreover, investigation comparing healthy young and older adults (i.e., no postural instability) would assist in determining if the decreases in head stability evident on the compliant surfaces are age-related.

Although a significant increase in head stabilization on the firm, low-friction surface was also not supported in the statistical analyses, older adults when challenged with the task of walking across the firm, low-friction surface did appear to minimize the degree of variability associated with head motions (i.e., RMS head acceleration values on the firm, low-friction surface were 15% lower than those observed on the firm, high-friction surface). Study results, however, may have been influenced by small group sample sizes. Specifically, the calculated observed power to detect a significant difference (i.e., alpha level set at $p \leq .05$) in the firm, low-friction and firm, high-friction comparison was .49. Consequently, an increase in the number of study participants may have provided the necessary statistical power to confirm the proposed hypothesis.

Several researchers have recently suggested that head stabilization serves to maintain a stable orientation of the visual and vestibular systems during task performance (Assaiante & Amblard, 1990; DiFabio & Emasithi, 1997; Holt et al., 1995; Pozzo et al., 1992). As such, head stability increases the perceptual sensitivity of the visual and vestibular systems to alterations in body kinematics associated with change in the individual-environment dynamics (e.g., upper-body perturbations due to slips) (Holt et al., 1995; Pozzo et al., 1992). Thus, the observed increase in head stability when older adults walked across the firm, low-friction surface may suggest that study participants attempted to maintain the effective use of visual and vestibular information in the perception and control of balance by increasing head stability.

Observed differences in the degree of head stability between the two groups of older adults during the gait cycle supported the hypothesis that older adults identified as at-high risk for falls would demonstrate significantly greater head stability across the three support surface conditions. This result is consistent with the previous findings of Clark and Dutto (1997), who demonstrated that when walking at a preferred speed across a firm, high-friction surface, older adults who self-reported a history of falls exhibited increased head stability when compared to a sample of healthy older adults (e.g., no previous history of falls within past 2 years). The increase in the degree of head stability when walking may be attributed to adaptations in head control exhibited by older adults at-risk for falls in order to maintain the effective use of visual and vestibular information in the perception and control of balance.

Coordination of lower extremity movement patterns

Significant differences in lower extremity joint kinematics evidenced across the three support surface conditions provided support for the hypothesis that variations in support surface properties would result in observable differences in the coordination and control of lower extremity movement patterns. These findings support those of Eng (1992) and O'Connor (1991), who also demonstrated changes in lower extremity kinematics when younger adult participants were required to walk onto and across a slippery support surface (i.e., firm, low-friction). Andres and O'Connor (1990) suggested that the observed variation in lower extremity joint kinematics when young adults walked across a firm, low-friction surface represent functional adaptations necessary for the achievement of the task goal (i.e., successful locomotion across the surface). As such, the variations in lower extremity coordination patterns evident across surface conditions in the present investigation may indicate that older adults, irrespective of level of postural stability, adapted coordination patterns to the collective constraints associated with the task goal and/or support surface dynamics.

Results further indicated that older adult participants were able to alter the relative foot angle at foot-surface contact to the constraints imposed by the dynamics of the firm, low-friction support surface. More specifically, when walking across the firm, low-friction surface, participants reduced the angle between the plantar surface of the foot and

the support surface at foot contact (i.e., participants had a flatter foot when contacting the surface). Previous researchers (Andres & O'Connor, 1990; Eng, 1992) have reported similar adaptations in the coordination and control of the foot, when study participants stepped onto or walked across slippery support surface conditions. These authors suggested that the observed reductions in the relative angle between the foot and surface at contact were associated with a decrease in the shear component of the ground reaction force. Thus, older adults in the present study may have adapted the coordination of the foot at surface contact to effectively minimize the shear or "slip" forces associated with heel strike in order to reduce the risk for slips and/or falls on a firm, low-friction surface.

Several previously reported investigations examining the gait patterns of older adults (Murray et al., 1969; Strandberg, 1983; Winter et al., 1990) demonstrated that reductions in the relative angle between the foot and the support surface at foot contact (i.e., flatter foot landing) were associated with shortened step lengths. Examination of the gait parameter data in the present study indicated that older adults did indeed decrease their step length (i.e., compared with firm, high-friction and compliant surfaces) when walking across the firm, low-friction support surface ($F(2,28) = 176.80$, $p \leq .001$). Although these results appear to provide support for the findings of Winter and colleagues (i.e., Winter, Patla, Frank & Walt, 1990), the relative foot angle at contact was not significantly correlated with step length across any of the three surface conditions. These findings may suggest that in the present investigation relative foot angle at contact was not constrained by step length (i.e., gait parameters) but instead constituted a functional adaptation in the coordination of movement to the imposed support surface constraint (i.e., firm, low-friction surface).

Control of lower extremity movement patterns

The hypothesis that variations in support surface properties would result in significant changes in the control of lower extremity movement patterns during the gait cycle was supported by results of the 2 x 3 x 2 (Group x Surface Condition x Trial Block) RM ANOVA. Findings indicated that older adult participants in the present study attempted to minimize the level of control complexity (i.e., reduced the level of activity) associated with lower extremity joint motions when walking on the firm, low-friction

surface. These results suggest that the lower extremity joints, especially the knee joint, were relatively fixed or rigid during the gait cycle. Rigidly fixed joint angles (i.e., freezing of the individual degrees of freedom) provided older adult participants with the opportunity to coordinate and effectively control the foot, shank and thigh segments as a "single", semi-rigid segment. This observed fixation or "freezing" of the degrees of freedom associated with the lower extremity joint motions during the gait cycle are similar to those reported for novel task acquisition (Vereijken et al., 1992) and the development of phylogenetic skills (Newell, Kugler, van Emmerik, & McDonald, 1989). Vereijken et al. (1992) suggested, "In solving a new and unfamiliar motor problem, the novice is required to reorganize the control of an overwhelming number of degrees of freedom in such a way as to be able to perform the task" (p.133). Thus, in the present investigation, participants appear to have coordinated the biomechanical degrees of freedom associated with the lower extremity joints in such a manner as to reduce the control complexity associated with gait patterns utilized for walking on a firm, low-friction support surface.

In contrast to the firm, low-friction surface condition, study participants exhibited an increase in the level of activity (i.e., a greater degree of control complexity) in the lower extremity joints when walking across the compliant support surface. That is, as opposed to maintaining a rigid, fixation of the degrees of freedom associated with the hip, knee and ankle joints, a release or freeing of degrees of freedom and subsequent increase in joint angle activity was evidenced on the compliant surface. This release of rigid control may be associated with the participants perceiving and exploiting the dynamics of the support surface properties. More specifically, when stepping on the compliant surface, the compressive forces associated with foot contact and the stance phases of the gait cycle produced a deformation of the open cell foam. This change in the structure of the foam not only "cushions" the foot during footfall but also results in the stance foot sinking "below" the top of the surface during the single support phase. Consequently, participants needed to actively alter joint kinematics to maintain toe clearance of the swing limb and produce adequate flexion in the stance limb to ensure toe off without stumbling.

Gait Parameters

Head motions and lower extremity kinematics exhibited by participants in the present investigation may have been influenced by gait parameters (e.g., gait velocity, stride length, and stride frequency) associated with self-selected walking speeds. Relationships between both head stability and lower extremity kinematics and gait parameters have been previously reported (Holt et al., 1995; Winter et al., 1990). For example, Winter and colleagues (Winter et al., 1990) demonstrated declines in the relative foot angle at surface contact with declines in gait velocity. Additionally, Holt and colleagues (Holt et al., 1995) report diminished head stability with increases in gait velocity and stride frequency.

Potential confounding effects of gait parameters on head motion and lower extremity kinematics in the present investigation were examined using both correlation matrices and RM ANOVAs. Pearson product moment correlation matrices were constructed using the variables; gait velocity, stride length, stride frequency, relative foot angle at surface contract, RMS head accelerations, and joint angular position standard deviations for the ankle, knee and hip across each surface condition and trial block (for a full compliment of correlation matrices see Appendix I). As indicated in the correlation matrices, stride length had a significant positive correlation with measurements of the control of ankle and knee joint motions and head stability across all surface conditions. The significant correlations between stride length and head stability on the firm, high-friction and compliant support surface conditions were observed for Trial Block II only. In contrast, significant correlations for these measures were observed for both Trial Block I and Trial Block II on the firm, low-friction support surface. Significant positive correlations were also observed between gait velocity and RMS head accelerations for both trial blocks performed on the firm, low-friction support surface. Gait velocity was not significantly correlated with RMS head acceleration when participants walked across the firm, high-friction or compliant surfaces. In regards to lower extremity joint control, significant positive correlations were observed between gait velocity and the control at the ankle joint for the compliant surface only. Significant correlations between gait velocity and knee joint control were evidenced both on the compliant surface and the firm, low-friction surface. Stride length and the control of joint motions for the ankle and

knee joints were significantly correlated across all surface conditions. Whereas, stride length and joint control at the hip had a significant positive correlation for the firm, low-friction surface only. In contrast to measures of head stability and the control of lower extremity joint motions, no significant correlations were determined for relative foot angle at surface contact and any gait parameter.

Separate univariate RM ANOVAs (Group x Surface Condition x Trial Block) conducted for the gait parameters; gait velocity, stride length and stride frequency indicated that each variable changed significantly across both Surface Condition and Trial Block (see Appendix J for descriptive statistics and Appendix K for RM ANOVA output tables, power estimates, and eta-squared values). Results also indicated that healthy older adults walked significantly faster and with longer step lengths as compared to older adults at-high risk for falls (see Appendix J and K).

Given the collective findings of both significant positive correlations between gait parameters (i.e., stride length and gait velocity) and measures of head stability and lower extremity joint control along with the significant differences between the two older adult groups on these measures, findings of the present investigation need to be interpreted with caution. The significant differences in gait velocity, stride length, and stride frequency across both surface conditions and trial blocks make it difficult to parse out potential confounding effects of these gait parameters on the head motions and lower extremity kinematics observed in the present investigation. Thus, additional investigations that examine the interaction between variations in gait parameters (i.e., task-goals) and surface dynamics during gait may be fruitful.

Conclusions

Results of the present investigation indicate that gait patterns adopted by older adults were influenced by variations in support surface properties. Older adults, irrespective of level of postural stability, altered the degree of stability in head trajectories as well as the coordination and control of lower extremity movement patterns during the gait cycle to accommodate changes in the friction and compliance of the support surface. Kinematic analyses yielded similar coordination patterns in the relative motions of the hip and knee during the gait cycle as well as the relative angle of the foot at surface for

the healthy and at-high risk for falls older adult groups. Although similar coordination patterns in the lower extremities emerged for the two groups of older adults, the degree of control of lower extremity joint motions as well as the degree of stability in head motions differed with respect to the participants' level of postural stability. The observed changes in the coordination and control of lower extremity joint motions and the degree of head stability in response to variations support surface constraints evidenced in the present investigation provides empirical support for the theory of perception and the control of bodily orientation (Riccio & Stoffregen, 1988). Specifically, findings indicate that during the goal-directed behavior of walking, the coordination and control of body segments emerge from the constraints imposed by the interaction of the support surface, the task, and the individual.

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CONCLUSION

Many tasks and activities associated with daily living require coordination and control of postures on a wide variety of support surfaces. As such, individuals often adapt movements and actions to accommodate these new demands almost immediately and unknowingly. This ability to alter and adapt coordination and control of body kinematics to accommodate variations in support surface dynamics generally occurs unnoticed until it is compromised. Previous research has suggested that for many older adults, especially those at-high risk for falls, age-associated changes in the perceptual (e.g., visual, vestibular, and somatosensory) and action (e.g., musculoskeletal) systems make it difficult to coordinate and control posture in response to changes in the support surface properties. Since falls may result from an inability of older adults to appropriately perceive and adapt to the environment, the study of the coordination and control of posture while performing tasks in different environments (i.e., support surfaces) may provide further insights into fall risk in this population.

The present studies were conducted to examine the influence of individual, task and environment constraints on the coordination and control of posture. The primary objective of the studies was to determine whether the coordination and control of body kinematics exhibited by older adults during stance (i.e., leaning forward through the region of stability) and gait would be different across variations in support surface properties. Secondary study objectives included, examining if coordination and control measures of body kinematics differed as a function of the participants' level of postural stability and/or repeated exposure to the support surface properties. The major finding common to the two studies was the influence of support surface dynamics on the coordination of movement. Significant differences in postural control strategies were evidenced across the different support surface conditions during the forward leaning task. Additionally, older adult participants exhibited changes in lower extremity coordination patterns during the gait cycle in response to variations in surface dynamics. Significant differences in coordination measures across variations in support surface conditions during the two different goal-directed tasks (leaning and walking) indicated that the body kinematics exhibited by both groups of older adults were influenced by the collective constraints of task and support surface dynamics.

The present findings further support the premise that surface properties constrain action. Significant differences in the measures of body kinematics in relation to the imposed support surface constraints provides evidence for the idea that "different forms of motion are uniquely related to the dynamics of each surface" (Stoffregen & Flynn, 1994; p36). During the gait cycle, lower extremity coordination patterns as evidenced in the angle-angle diagrams demonstrated distinguishing surface-specific patterns as participants walked across surfaces with different support properties. In addition, qualitative observation of the postural control strategies utilized by participants when performing the forward leaning task also indicated that strategies were, in general, surface-appropriate.

Although it was hypothesized that the level of postural stability would influence body kinematics (e.g., head stability, postural control strategies, lower extremity coordination), differences between groups emerged only in the gait study. This finding may indicate that task constraints become most evident when older adults are engaged in more posturally demanding tasks. Collectively, the present findings indicate that in the present investigation body kinematics were constrained by the support surface properties. The present results provide further support for the theory of perception and control of bodily orientation and its central premise that coordinated actions emerge from the constraints imposed by the individual, task and environment.

Much of the research examining changes in the postural control of older adults has neglected the environment as it influences the coordination and control of body kinematics. The present findings would suggest that the constraints imposed by the support surface undoubtedly influence goal-directed behaviors (i.e., forward lean and gait). Future investigations should seek to examine the constraints imposed by variations in the properties of a single surface, as well as testing across a wider variety of support surface conditions. Although task constraints were addressed in the present investigation, the task utilized were two distinct tasks with different task-goals. Future investigations should seek to examine variations in task goals within a single task context (e.g., walking at different speeds, leaning to different regions within the limits of stability).

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APPENDICES

APPENDIX A

Background and Medical History Questionnaire

Background and Medical History Questionnaire

Name _____ Birthdate_ _____

Address _____

_____

Phone _____

Do you require eyeglasses? **YES/NO** hearing aid? **YES/NO**
Are you currently taking **any** prescription medications? **YES/NO**

If you answer **YES**, please list those medications and the condition for which they are prescribed.

Do you currently have any medical condition for which you see a physician regularly? **YES/NO**
If you answer **YES**, please describe the condition and the treating physician's name.

Have you required emergency medical care OR hospitalization in the last **three years? YES/NO**
If you answered **YES**, please list when this occurred and briefly explain why.

Have you ever had any condition or suffered any injury that has affected your balance or ability to walk without assistance? **YES/NO**
If your answered **YES**, please describe the condition/injury and relevant history.

Have you experienced a fall within the last 3 years? **YES/NO** # of falls:
Did you require medical treatment? **YES/NO**
If you answered **YES** to either question, please list the approximate **date of the fall**, the **medical treatment required**, and the **reason you fell** in each case (e.g. uneven surface, going down stairs, etc.).

How would you describe you balance **(circle one)**:
Excellent / Good / Average / Fair / Poor

Have you ever been diagnosed as having any of the following conditions?

Cancer (if yes, describe what kind)	YES / NO
Heart problems / Angina / High blood pressure	YES / NO
Peripheral vascular disease	YES / NO
Chemical dependency (alcoholism)	YES / NO
Diabetes	YES / NO
Neuropathies (problem with sensation	YES / NO
Multiple Sclerosis	YES / NO
Movement disorders	YES / NO
Rheumatoid arthritis / Other arthritic conditions	YES / NO
Depression	YES / NO
Stroke / Head Injury	YES / NO
Epilepsy / Seizures	YES / NO
Neurological problems	YES / NO
Parkinson's disease	YES / NO
Cerebellar problems (ataxia)	YES / NO
Transient ischemic attacks	YES / NO
Polio / Post polio syndrome	YES / NO
Inner ear problems / Recurrent ear infections	YES / NO
Visual/Depth perception problems	YES / NO

Do you suffer any of the following symptoms in your legs or feet?

Numbness / Loss of feeling	YES / NO
Tingling	YES / NO
Arthritis	YES / NO
Swelling	YES / NO

APPENDIX B

Berg Balance Scale

Berg Balance Scale

1. Sitting to Standing
   Instructions: Please stand up. Try not to use your hands for support
   ( )4 able to stand without using hands and stabilize independently
   ( )3 able to stand independently using hands
   ( )2 able to stand using hands after several tries
   ( )1 needs minimal aid to stand or to stabilize
   ( )0 needs moderate or maximal assist to stand

2. Standing Unsupported
   Instructions: Please stand for two minutes without holding.
   ( )4 able to stand safely 2 minutes
   ( )3 able to stand 2 minutes with supervision
   ( )2 able to stand 30 seconds unsupported
   ( )1 needs several tries to stand 30 seconds unsupported
   ( )0 unable to stand 30 seconds unassisted

If a subject is able to stand 2 minutes unsupported, score full points for sitting unsupported. Proceed to item #4.

3. Sitting with back unsupported but feet supported on floor or on a stool
   Instructions: Please sit with arms folded for 2 minutes
   ( )4 able to sit safely and securely 2 minutes
   ( )3 able to sit 2 minutes under supervision
   ( )2 able to sit 30 seconds
   ( )1 able to sit 10 seconds
   ( )0 able to sit without support 10 seconds

4. Standing to sitting.
   Instruction: Please sit down
   ( )4 sits safely with minimal use of hands
   ( )3 controls descent by using hands
   ( )2 uses back of legs against chair to control descent
   ( )1 sits independently but has uncontrolled descent
   ( )0 needs assistance to sit

5. Transfers
   Instructions: Arrange chair(s) for a pivot transfer. Ask subject to transfer one way toward a seat with armrests and one way toward a seat without armrests. Your may use low chairs (one with and one without armrests) or a bed and a chair.
   ( )4 able to transfer safely with minor use of hands
   ( )3 able to transfer safely definite need of hands
   ( )2 able to transfer with verbal cuing and/or supervision
   ( )1 needs one person to assist
   ( )0 needs two people to assist or supervisor to be safe

6. Standing unsupported with eyes closed
Instructions: Please close your eyes and stand still for 10 seconds.
( )4 able to stand 10 seconds safely
( )3 able to stand 10 seconds with supervision
( )2 able to stand 3 seconds
( )1 unable to keep eyes closed 3 seconds but stays safely
( )0 needs help to keep from falling

7. Standing unsupported with feet together
Instructions: Place your feet together and stand without holding
( )4 able to place feet together independently and stand 1 minute safely
( )3 able to place feet together independently and stand for 1 minute with supervision
( )2 able to place feet together but unable to hold for 30 seconds
( )1 needs help to attain position but able to stand 15 seconds feet together
( )0 needs help to attain position and unable to hold for 15 seconds

8. Reaching forward with outstretched arm while standing
Instructions: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can (examiner placed a ruler at end of fingertips when arms is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance forward that the finger reach while the subject is in the most forward lean position. When possible, ask subject to use both arms when reaching to avoid rotation of the trunk.)
( )4 can reach forward confidently >25 cm (10 inches)
( )3 can reach forward >12 cm safely (5 inches)
( )2 can reach forward >5 cm safely (2 inches)
( )1 reaches forward but needs supervision
( )0 loses balance while trying/requires external support

9. Pick up object from the floor from a standing position.
Instructions: Pick up the shoe/slipper which is placed in front of your feet.
( )4 able to pick up slipper safely and easily
( )3 able to pick up slipper but needs supervision
( )2 unable to pick up but reaches 2-5 cm (1-2 inches) from slipper and keeps balance independently
( )1 unable to pick up and needs supervision trying
( )0 unable to try/needs assist to keep from losing balance or falling

10. Turning to look behind over left and right shoulders while standing
Instruction: Turn to look directly behind you over toward left shoulder. Repeat to the right. Examiner may pick an object to look at directly behind the subject to encourage a better twist turn.
( )4 looks behind from both sides and weight shifts well
( )3 looks behind one side only other side shows less weight shift
( )2 turns sideways only but maintains balance
( )1 needs supervision when turning
( )0 needs assistance while turning

11. Turn 360 Degrees
Instructions: Turn completely around in a full circle. Pause. Then turn a full circle in the other direction.
( )4 able to turn 360 degrees safely in 4 seconds or less
( )3 able to turn 360 degrees safely in only 4 seconds or less
( )2 able to turn 360 degrees safely but slowly
( )1 needs close supervision or verbal cuing
( )0 needs assistance while turning

12. Place alternate foot on step or stool while standing unsupported
Instructions: Place each foot alternately on the step/stool. Continue until foot has touched the step/stool four times.
( )4 able to stand independently and safely and complete 8 steps in 20 seconds
( )3 able to stand independently and complete 8 steps in 20 seconds
( )2 able to complete 4 steps without aid with supervision
( )1 able to complete >2 steps needs minimal assist
( )0 needs assistance to keep from falling/unable to try

13. Standing unsupported one foot in front
Instructions: (Demonstrate to Subject) Place one foot directly in front of the other. If you feel that you cannot place your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot. (To score 3 points, the length of the step should exceed the length of the other foot and the width of the stance should approximate the subject's normal stride width.
( )4 able to place foot tandem independently and hold 30 seconds
( )3 able to place foot ahead of other independently and hold 30 seconds
( )2 able to take small step independently and hold 30 seconds
( )1 needs help to step but can hold 15 seconds
( )0 loses balance while stepping or standing

14. Standing on one leg
Instructions: Stand on one leg as long as you can without holding.
( )4 able to lift leg independently and hold >10 seconds
( )3 able to lift leg independently and hold 5-10 seconds
( )2 able to lift leg independently and hold = or >3 seconds
( )1 tries to lift leg unable to hold 3 seconds but remains standing independently
( )0 unable to try or needs assist to prevent fall

( ) Total Score (Maximum - 56)

APPENDIX C

Informed Consent Document

TITLE: Task and Support Surface Constraints on the Coordination and Control of Posture in Older Adults

PRINCIPLE INVESTIGATOR: Dr. Debra Rose (Sean Clark)

PURPOSE: The purpose of this study is to examine the coordination and control of balance while posturally stable and unstable older adults **lean forward as far as possible while standing** and walk on different support surfaces. We are particularly interested in how older adults change their balance strategies to maintain a stable, upright position on surfaces that are rigid and slip-resistant, soft and pliable, and rigid and slippery.

I have received a verbal explanation of the study and:

I am aware that I will be participating in a pre-study screening on two occasions on two consecutive days during which my balance and mobility will be assessed.

I also recognize that as a result of my performance on these assessments, I may or may not be a suitable candidate for this study and thus, may or may not be asked to participate in the study.

I am aware that if I am a suitable candidate and I am willing to participate, I will be participating in two addition evaluations conducted one week apart. Each session will last approximately 1 hour and will consist of assessments related to my ability to balance and stay upright. During the first evaluation, my control of balance will be assessed while I stand quietly **then lean forward as far as I can safely without losing my balance. I will be asked to lean forward while standing** on each of three test surfaces. I will be asked to complete 12 trials on each of the three surfaces for a total of 36 trials. The surface conditions used during this testing will be: (1) rigid and slip-resistant, (2) soft and pliable, and (3) rigid and slippery. I will be permitted to rest after every **third** trial and between sets of trials on a new surface. The assessment during the second week will consist of being evaluated while walking, at my most comfortable speed, a distance of approximately 6 yards (~20 feet) across the same three surfaces. I will be asked to walk this distance 20 times on each of the three different surface types for a total of 60 trials. I will be provided with the opportunity to rest between each trial and between each set of trials on a new surface. For all testing, I will be required to walk and/or stand barefoot.

I understand that the potential benefits for me resulting from my involvement in this investigation include the opportunity to develop a better understanding of my ability to stand/**lean forward** and walk on a wide variety of different support surfaces. Also, the findings may help me to better identify the types of surfaces most likely to pose

the greatest risk for instability (e.g., wet tile surfaces, wet pavement, uneven grassy or sandy terrain's). Additionally, results of this study will contribute to the general understanding of balance control and the development of intervention programs aimed at improving balance in older adults.

I am also aware that the potential risks to me as a participant involved in this investigation are considered minimal. The use of an overhead harness system should reduce my anxiety levels during the assessment and will prevent me from falling during the testing procedures. An assistant will also stand and/or walk close to me during each of the tests performed. I also recognize that I may experience muscle soreness or discomfort as a result of the additional effort I may need to exert when maintaining balance or walking on "unfamiliar" surfaces. This muscular discomfort should, however, be a short-term effect with no lasting implications.

I am aware that my confidentiality will be maintained at all times throughout the study by using a personalized identification number. Also, at no time will my name appear on record forms, computer files, publications or presentations related to the study.

I understand that the University does not provide a research participant with compensation or medical treatment in the event I am injured as a result of participation in the study.

I have been informed about the nature of the study and understand why it is being conducted. The researchers have provided me with an opportunity to ask further questions about any aspect of this study. I understand that my participation is voluntary, that I may decide at any time not to participate in this study, and that I may withdraw my consent at any time without prejudice to my relations with Oregon State University. Any questions that I have about the research or any aspect of my participation should be directed to Dr. Debra Rose or Sean Clark at 737-5934.

Questions concerning my rights as a participant in this research can be directed to the Institutional Review Board for the Protection of Human Subjects at the Research Office of Oregon State University (737-3437).

I have received a copy of this consent form for my records.

_____	_____
Signature of Participant	Name of Participant
_____	
Date Signed	
_____	_____
Signature of Principal Investigator (optional)	Date Signed

APPENDIX D

Stance Study: Cross-correlation coefficient
RM ANOVA output table, observed power, and eta-squared values

Stance Study: Cross-correlation RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	3.243	1	3.243	1.158	.300	.076	.171
Surface Condition	3.560	2	1.780	12.784	.000	.477	.993
Surface Condition* Group	5.731E-04	2	2.866E-04	.002	.998	.000	.050
Error (Surface Condition)	3.899	28	.139				
Trial Block	.247	1	.247	2.105	.169	.131	.272
Trial Block * Group	.148	1	.148	1.260	.280	.083	.182
Error (Trial Block)	1.640	14	.117				
Surface Condition * Trial Block	2.333E-02	2	1.166E-02	.153	.859	.011	.071
Surface Condition * Trial Block * Group	.191	2	9.535E-02	1.251	.302	.082	.249
Error (Surface Condition * Trial Block)	2.135	28	7.625E-02				

Computed using alpha = .05

APPENDIX E

Stance Study: RMS head acceleration RM ANOVA output table, observed power, and eta-squared values

Stance Study: RMS head acceleration RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	10935.882	1	10935.882	.348	.565	.024	.085
Surface Condition	34583.864	2	17291.932	12.120	.000	.464	.991
Surface Condition* Group	5298.207	2	2649.103	1.857	.175	.117	.354
Error (Surface Condition)	39948.785	28	1426.742				
Trial Block	100.103	1	100.103	.117	.737	.008	.062
Trial Block * Group	1330.762	1	1330.762	1.558	.232	.100	.214
Error (Trial Block)	11956.386	14	854.028				
Surface Condition * Trial Block	7427.483	2	3713.742	6.257	.006	.309	.860
Surface Condition * Trial Block * Group	488.198	2	244.099	.411	.667	.029	.110
Error (Surface Condition * Trial Block)	16618.214	28	593.508				

Computed using alpha = .05

APPENDIX F

Gait Study: RMS head acceleration RM ANOVA output table, observed power, and eta-squared values

Gait Study: RMS head acceleration RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	451778.7	1	451778.7	5.025	.042	.264	.550
Surface Condition	271231.793	2	135615.896	5.450	.010	.280	.806
Surface Condition* Group	54419.283	2	27209.642	1.093	.349	.072	.222
Error (Surface Condition)	696765.967	28	24884.499				
Trial Block	710.206	1	710.206	.211	.653	.015	.071
Trial Block * Group	1558.702	1	1558.702	.464	.507	.032	.097
Error (Trial Block)	47058.539	14	3361.324				
Surface Condition * Trial Block	19.917	2	9.959	.003	.997	.000	.050
Surface Condition * Trial Block * Group	6219.421	2	3109.710	.908	.415	.061	.191
Error (Surface Condition * Trial Block)	95938.828	28	3426.387				

Computed using alpha = .05

APPENDIX G

Gait Study: Standard deviations of joint angular positions
RM ANOVA output table, observed power, and eta-squared values

Gait Study: Standard deviations of joint angular positions
RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	215.941	1	215.941	15.082	.002	.519	.950
Error	200.445	14	14.317				
Joint	6261.386	2	3130.693	573.927	.000	.976	1.000
Joint * Group	7.751	2	3.876	.710	.500	.048	.158
Error (Joint)	152.736	28	5.455				
Surface Condition	3236.086	2	1618.043	390.696	.000	.965	1.000
Surface Condition * Group	4.341	2	2.171	.524	.598	.036	.128
Error (Surface Condition)	115.960	28	4.141				
Trial Block	.893	1	.893	.826	.379	.056	.136
Trial Block* Group	.689	1	.689	.638	.438	.044	.116
Error (Trial Block)	15.133	14	1.081				
Joint * Surface Condition	1263.535	4	315.884	156.223	.000	.918	1.000
Joint * Surface Condition * Group	6.322	4	1.580	.782	.542	.053	.235
Error (Joint * Surface Condition)	113.232	56	2.022				
Joint * Trial Block	2.596	2	1.298	3.757	.036	.212	.638
Joint * Trial Block * Group	.256	2	.128	.370	.694	.026	.104
Error (Joint * Trial Block)	9.676	28	.346				
Surface Conditon * Trail Block	1.016	2	.508	.693	.508	.047	.155
Surface Conditon * Trail Block * Group	1.418	2	.709	.967	.393	.065	.201
Error (Surface Conditon * Trail Block)	20.532	28	.733				
Joint * Surface Conditon * Trail Block	2.472	4	.618	1.781	.145	.113	.509
Joint * Surface Conditon * Trail Block * Group	.315	4	7.877E-02	.227	.922	.016	.096
Error (Joint * Surface Conditon * Trail Block)	19.434	56	.347				

Computed using alpha = .05

APPENDIX H

Gait Study: Relative foot angle at surface contact
RM ANOVA output table, observed power, and eta-squared values

Gait Study: Relative foot angle at surface contact
RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	Df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	167.910	1	167.910	.717	.411	.049	.124
Surface Condition	757.373	2	378.687	31.099	.000	.690	1.000
Surface Condition* Group	5.505	2	2.753	.226	.799	.016	.082
Error (Surface Condition)	340.945	28	12.177				
Trial Block	2.972E-02	1	2.972E-02	.018	.894	.001	.052
Trial Block * Group	1.790	1	1.790	1.104	.311	.073	.165
Error (Trial Block)	22.697	14	1.621				
Surface Condition * Trial Block	11.002	2	5.501	4.594	.019	.247	.731
Surface Condition * Trial Block * Group	1.628	2	.814	.680	.515	.046	.153
Error (Surface Condition * Trial Block)	33.528	28	1.197				

Computed using alpha = .05

APPENDIX I

Gait Study: Correlation matrices for gait parameters and dependent variables

Gait Study: correlation matrix for the Firm, high-friction surface condition, Trial Block I.

	Gait Velocity	Stride Length	Stride Frequency	RMS head acceleration	Relative foot angle	Ankle joint control	Knee joint control	Hip joint control
Gait Velocity	1.00							
Stride Length	.820*	1.00						
Stride Frequency	.688*	.164	1.00					
RMS head acceleration	-.023	.428	-.553*	1.00				
Relative foot angle	-.181	-.201	-.023	-.203	1.00			
Ankle Joint Control	.447	.612*	.054	.182	-.272	1.00		
Knee Joint Control	.520*	.605*	.179	-.047	-.001	.845*	1.00	
Hip Joint Control	.024	.062	-.109	.028	.195	-.076	-.132	1.00

* Significant Pearson correlation ($p \leq .05$)

Gait Study: correlation matrix for the Firm, high-friction surface condition, Trial Block II.

	Gait Velocity	Stride Length	Stride Frequency	RMS head acceleration	Relative foot angle	Ankle joint control	Knee joint control	Hip joint Control
Gait Velocity	1.00							
Stride Length	.824*	1.00						
Stride Frequency	.643*	.112	1.00					
RMS head acceleration	.201	.594*	-.392	1.00				
Relative foot angle	.041	-.084	.196	-.146	1.00			
Ankle Joint Control	.453	.605*	.046	.300	-.219	1.00		
Knee Joint Control	.465	.524*	.148	.057	-.095	.706*	1.00	
Hip Joint Control	.366	.340	.095	.372	.107	.197	.115	1.00

* Significant Pearson correlation ($p \leq .05$)

Gait Study: correlation matrix for the Compliant surface condition, Trial Block I.

	Gait Velocity	Stride Length	Stride Frequency	RMS head acceleration	Relative foot angle	Ankle joint control	Knee joint control	Hip joint Control
Gait Velocity	1.00							
Stride Length	.913*	1.00						
Stride Frequency	.762*	.454	1.00					
RMS head acceleration	.147	.413	-.291	1.00				
Relative foot angle	.102	.076	.021	-.219	1.00			
Ankle Joint Control	.741*	.778*	.389	.166	.095	1.00		
Knee Joint Control	.693*	.657*	.136	.141	-.026	.676*	1.00	
Hip Joint Control	.261	.426	-.063	.409	-.137	.346	.314	1.00

* Significant Pearson correlation ($p \leq .05$)

Gait Study: correlation matrix for the Compliant surface condition, Trial Block II.

	Gait Velocity	Stride Length	Stride Frequency	RMS head acceleration	Relative foot angle	Ankle joint control	Knee joint control	Hip joint Control
Gait Velocity	1.00							
Stride Length	.915*	1.00						
Stride Frequency	.502*	.123	1.00					
RMS head acceleration	.383	.552*	-.230	1.00				
Relative foot angle	.148	.032	.274	-.292	1.00			
Ankle Joint Control	.694*	.758*	.065	.202	.133	1.00		
Knee Joint Control	.571*	.437	.536*	-.196	.378	.491	1.00	
Hip Joint Control	.346	.343	.181	-.158	.072	.336	.497	1.00

* Significant Pearson correlation ($p \leq .05$)

Gait Study: correlation matrix for the Firm, low-friction surface condition, Trial Block I.

	Gait Velocity	Stride Length	Stride Frequency	RMS head acceleration	Relative foot angle	Ankle joint control	Knee joint control	Hip joint Control
Gait Velocity	1.00							
Stride Length	.806*	1.00						
Stride Frequency	.687*	.188	1.00					
RMS head acceleration	.883*	.771*	.439	1.00				
Relative foot angle	-.091	-.069	.139	-.281	1.00			
Ankle Joint Control	.474	.801*	-.036	.408	-.114	1.00		
Knee Joint Control	.586*	.762*	.171	.510*	.329	.574*	1.00	
Hip Joint Control	.556*	.843*	-.022	.525*	-.150	.739*	.659*	1.00

* Significant Pearson correlation ($p \leq .05$)

Gait Study: correlation matrix for the Firm, low-friction surface condition, Trial Block II.

	Gait Velocity	Stride Length	Stride Frequency	RMS head acceleration	Relative foot angle	Ankle joint control	Knee joint control	Hip joint Control
Gait Velocity	1.00							
Stride Length	.797*	1.00						
Stride Frequency	.556*	.004	1.00					
RMS head acceleration	.892*	.817*	.341	1.00				
Relative foot angle	-.130	-.191	.129	-.202	1.00			
Ankle Joint Control	.387	.825*	-.348	.418	-.038	1.00		
Knee Joint Control	.509*	.750*	-.090	.602*	.250	.653*	1.00	
Hip Joint Control	.378	.728*	-.287	.375	-.060	.827*	.536*	1.00

* Significant Pearson correlation ($p \leq .05$)

APPENDIX J

Gait Study: Gait Velocity, Stride Length, and Stride Frequency Means and Standard Deviations

Gait Study: Gait Velocity Means and Standard Deviations

		At-high risk for falls		Healthy	
Surface	Trial Block	Mean	StDev	Mean	StDev
Firm, high-friction	I	.776	.148	1.013	.145
	II	.852	.149	1.028	.157
Compliant	I	.584	.136	.809	.275
	II	.653	.122	.889	.135
Firm, low-friction	I	.181	.008	.311	.166
	II	.199	.005	.350	.189

Gait Study: Stride Length Means and Standard Deviations

		At-high risk for falls		Healthy	
Surface	Trial Block	Mean	StDev	Mean	StDev
Firm, high-friction	I	.945	.117	1.197	.085
	II	.975	.114	1.205	.107
Compliant	I	.785	.138	1.115	.231
	II	.876	.089	1.164	.157
Firm, low-friction	I	.353	.079	.604	.191
	II	.402	.080	.660	.202

Gait Study: Stride Frequency Means and Standard Deviations

		At-high risk for falls		Healthy	
Surface	Trial Block	Mean	StDev	Mean	StDev
Firm, high-friction	I	.824	.109	.853	.109
	II	.875	.103	.861	.098
Compliant	I	.706	.107	.717	.131
	II	758	.085	.774	.062
Firm, low-friction	I	.505	.168	.497	.175
	II	.520	.174	.524	.158

APPENDIX K

Gait Study: Gait Velocity, Stride Length, and Stride Frequency
RM ANOVA output tables, observed power, and eta-squared values

Gait Study: Gait Velocity RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	Df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	.946	1	.946	11.247	.005	.446	.877
Error	1.174	14	8.387E-02				
Surface Condition	7.304	2	3.652	151.827	.000	.916	1.000
Surface Condition * Group	4.764E-02	2	2.382E-02	.990	.384	.066	.205
Error (Surface Condition)	.674	28	2.405E-02				
Trial Block	7.352E-02	1	7.352E-02	18.139	.001	.564	.977
Trial Block * Group	2.752E-03	1	2.752E-03	.679	.424	.046	.120
Error (Trial Block)	5.674E-02	14	4.053E-03				
Surface Conditon * Trial Block	1.748E-02	2	8.739E-03	2.187	.131	.135	.409
Surface Condition * Trial Block * Group	6.615E-03	2	3.307E-03	.828	.447	.056	.177
Error (Surface Condition * Trail Block)	.112	28	3.996E-03				

Computed using alpha = .05

Gait Study: Stride Length RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	Df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	1.730	1	1.730	22.357	.000	.615	.992
Error	1.083	14	7.738E-02				
Surface Condition	6.102	2	3.051	176.795	.000	.927	1.000
Surface Condition * Group	2.063E-02	2	1.032E-02	.598	.557	.041	.139
Error (Surface Condition)	.483	28	1.726E-02				
Trial Block	5.357E-02	1	5.357E-02	26.211	.000	.652	.997
Trial Block * Group	2.125E-03	1	2.125E-03	1.040	.325	.069	.158
Error (Trial Block)	2.861E-02	14	2.044E-03				
Surface Conditon * Trial Block	1.087E-02	2	5.434E-03	1.737	.195	.110	.333
Surface Condition * Trial Block * Group	2.505E-03	2	1.252E-03	.400	.674	.028	.108
Error (Surface Condition * Trail Block)	8.761E-02	28	3.129E-03				

Computed using alpha = .05

Gait Study: Stride Frequency RM ANOVA output table, observed power, and eta-squared values

Source	Sum of Squares	Df	Mean Square	F	Sig.	Eta Squared	Observed Power
Group	9.437E-04	1	9.437E-04	.019	.893	.019	.052
Error	.711	14	5.075E-02				
Surface Condition	1.938	2	.969	48.608	.000	.776	1.000
Surface Condition * Group	1.011E-03	2	5.053E-04	.025	.975	.002	.053
Error (Surface Condition)	.558	28	1.994E-02				
Trial Block	2.958E-02	1	2.958E-02	9.797	.007	.412	.829
Trial Block * Group	4.378E-04	1	4.378E-04	.145	.709	.010	.065
Error (Trial Block)	4.226E-02	14	3.019E-03				
Surface Conditon * Trial Block	4.957E-03	2	2.478E-03	.968	.392	.065	.201
Surface Condition * Trial Block * Group	3.674E-03	2	1.837E-03	.718	.497	.049	.159
Error (Surface Condition * Trail Block)	7.168E-02	28	2.560E-03				

Computed using alpha = .05