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The biomechanics of slips are an important component in the prevention of fall-related injuries. The purpose of this paper is to review the available literature on the biomechanics of gait relevant to slips. This knowledge can be used to develop slip resistance testing methodologies and to determine critical differences in human behaviour between slips leading to recovery and those resulting in falls. Ground reaction forces at the shoe-floor interface have been extensively studied and are probably the most critical biomechanical factor in slips. The ratio of the shear to normal foot forces generated during gait, known as the required coefficient of friction (RCOF) during normal locomotion on dry surfaces or 'friction used/achievable' during slips, has been one biomechanical variable most closely associated with the measured frictional properties of the shoe/floor interface (usually the coefficient of friction or COF). Other biomechanical factors that also play an important role are the kinematics of the foot at heel contact and human responses to slipping perturbations, often evident in the moments generated at the lower extremity joints and postural adaptations. In addition, it must be realized that the biomechanics are dependent upon the capabilities of the postural control system, the mental set of the individual, and the perception of the environment, particularly, the danger of slipping. The focus of this paper is to review what is known regarding the kinematics and kinetics of walking on surfaces under a variety of environmental conditions. Finally, we discuss future biomechanical research needs to help to improve walkway-friction measurements and safety.

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1. Introduction

Pedestrian accidents on walkways continue to be a very serious problem. An analysis of data in 1986 found that the costs of pedestrian accidents was second in magnitude only to automobile accidents and that falls were the leading cause of accidental death in senior citizens (Rice *et al.* 1989). Pedestrian-fall accidents have been the second largest generator of unintentional workplace fatalities (Leamon and Murphy 1995), and accounted for nearly 11% and 20%, respectively, of all fatal and non-fatal (involving lost work days) occupational injuries in the USA in 1996 (US Department of Labor, Bureau of Labor Statistics 1997, 1998).

Falls precipitated by slipping are of major concern (Courtney *et al.* 2001). Lloyd and Stevenson (1992) reported that slips and trips cause 67% and 32% of falls sustained by the elderly and young, respectively. The magnitude of the problem is probably greater than suggested by the statistics above, as muscular strain or back pain illnesses resulting from slip and recovery incidents are usually not reported as related to slips (Manning and Shannon 1981, Troup *et al.* 1981, Anderson and Lagerlof 1983).

Understanding what causes slip-precipitated pedestrian accidents is challenging because of the multiple, interacting environmental and human factors involved. Among the environmental factors are properties of the walking-surface (such as surface roughness, compliance, topography, as well as the properties of adjacent areas and contaminants) and/or shoe or foot (e.g. material properties, tread and wear). Other environmental factors include lighting and contrast levels, and climatic factors such as ice and snow. Human factors include gait, expectation, the health of the sensory systems (i.e. vision, proprioception, somatosensation, and vestibular) and the health of the neuromuscular system.

One fundamental principal in determining the slip propensity of a given situation is the relationship between the friction required by the pedestrian for the manoeuvre being conducted (*Required friction*) compared with the friction available at the walkway/shoe interface (*Available friction*). Theoretically, as long as the available friction exceeds the required friction, the pedestrian will not slip. There are a variety of pedestrian gaits, e.g. level walking, load carrying, walking up ramps, that have different levels of required friction to prevent slip. Thus, biomechanical analysis of gait is potentially a valuable tool in the reduction of slip-induced fall accidents because it can illuminate the conditions that may be hazardous to pedestrians. Furthermore, biomechanical analysis of gait can be an important input into the setting of available friction thresholds to determine whether or not a shoe, walkway surface, or combination of the two will—or will not—be slip resistant (Marpet 1996).

The utility of biomechanics in the measurement of slipperiness goes beyond the matter of determining required friction. Biomechanics can be employed to 'tune' tribometric instruments (here, walkway-surface friction testing instruments) to reflect the friction situation facing pedestrians. The friction model taught in high-school and college physics classes (the Amontons-Coulomb model) assumes that friction is solely a material property of the interface materials, independent of contact area, pressure, temperature, velocity, etc. (Amontons 1699, Coulomb 1781). This model, true for friction between two rigid bodies, is an oversimplification with respect to pedestrian friction. James (1980, 1983) showed that the materials commonly used in shoe-bottom construction do not follow the assumptions of Amontons-Coulomb. Given that, devices that measure the frictional properties of the shoe-floor interface will give more meaningful results if testing conditions, such as pressure, velocity, contact time and so forth, mimic those of pedestrian gait. The need for such *biofidelity* in friction testing has been recognized for some time (Proctor and Coleman 1988). At present, no walkway tribometer has operational characteristics mimicking human gait. New methods do attempt to obtain biomechanically relevant measures of slip resistance (Grönqvist *et al.* 1989, Wilson 1990, Redfern and Bidanda 1994). Gait parameters describing foot dynamics during actual slip events under varying environmental conditions (floor and contaminant) are important in the further development of these devices.

Another aspect of slip-precipitated falls that incorporates biomechanics is the capability of the human postural control system that is used to maintain balance and recover from perturbations. Balance recovery during slips involves neuromuscular control, biomechanics, and their interaction with the environment. Therefore, to understand the impact of the environment on the potential of slips and falls, some knowledge of the human reactions to slip and the capability to recover from slip perturbations must be taken into consideration. Factors to consider in balance recovery and control are anatomical (e.g. foot geometry, body mass and its distribution, or segment length and height), physiological (e.g. strength, rate of muscle force rise, or gains and delays of feedback control), or perhaps cognitive and behavioural constraints (e.g. reaction time, attention, or fear of falling). Each of these constraints has a different impact on the ability to recover balance, which can be assessed in terms of kinematics and kinetics of the performance.

Thus, the main goal of this paper is to review the relevant literature investigating gait biomechanics and postural control while walking on surfaces of varying slipperiness. The focus of this literature review will be on the kinematics and kinetics of walking on surfaces of different inclination under a variety of environmental conditions. A second goal is to define the specific research directions and goals needed to determine which gait parameters are important in supporting improved walkway-friction measurements and safety.

2. Biomechanics of locomotion without slipping

This section describes the biomechanics of common locomotor activities without slipping. In all cases, the biomechanical descriptions include ground reaction forces, kinematics and moments generated at the lower extremity joints. Normal gait is first described for walking on a level surface, and then on inclined surfaces. The figures for level walking and inclined surface walking (when available) have been combined for ease of comparison (figures 1, 3–6). The next section describes the biomechanics of another locomotor activity known to be associated with slips and falls, namely, ascending and descending stairs. Finally, the influence of load carrying during walking on the biomechanics of gait is considered.

2.1. *Walking on level surfaces*

2.1.1. *Ground reaction forces*: The force interactions between the shoe and floor are probably the most critical biomechanical parameters in slips and falls. If the shear forces generated during a particular step exceed the frictional capabilities of the shoe/floor interface, then a slip is inevitable. Thus, an understanding of the forces at the shoe/floor interface is important. A number of researchers have examined foot forces, often termed ground reaction forces (GRF), during normal gait on a level surface (Strandberg and Lanshammar 1981, Perkins and Wilson 1983, Strandberg 1983, Winter 1991, Redfern and Dipasquale 1997, Cham and Redfern 2001a). (See

Table 1. Foot force parameters for normal walking on level surfaces (0°) and inclined surfaces (5° and 10°).

Variable: Mean (SD)	0°	5°	10°
First peak of normal forces (body weight N kg^{-1})	10.92 (1.42)†	12.15 (1.41)†	13.33 (1.52)†
Peak shear forces (body weight N kg^{-1})	1.77 (0.61)†	2.94 (0.56)†	4.06 (0.81)†
Timing of peak RCOF (% stance or ms)	16.5 (2.4)†% 91 (25)††ms	18.1 (3.6)†	19.2 (4.6)†
Timing of first peak of normal forces (% stance)	24.5 (5.2)†	21.4 (4.3)†	18.6 (5.4)†
Timing of peak shear forces (% stance)	19.0 (3.1)†	19.5 (2.6)†	19.0 (4.6)†
Peak RCOF*	0.17 (0.04)†† 0.18 (0.05)† 0.18 (0.06)‡ 0.20¶ 0.22§	0.26 (0.03)†	0.32 (0.05)† 0.33 (0.04)‡

*Required Coefficient of Friction, defined in section 2.2.1.
From †Cham and Redfern [vinyl floors] (2001a), ‡Hanson *et al.* (1999), §Perkins (1978), ¶Redfern and DiPasquale (1977), ††Strandberg (1983) ['grip' trials].

table 1 for a list of critical GRF parameters.) The normal forces (perpendicular to the walking surface) are typically characterized by two peaks (solid line in figure 1). The first peak occurs at the end of the loading phase (about 25% into stance) as full body weight is transferred to the supporting foot, while the second peak occurs later in stance just prior to the beginning of the toe-off phase. The anterior-posterior shear forces exhibit a biphasic, symmetrical shape with the first major peak in the forward direction attributed to the loading dynamics, while the second maximum in the rearward direction happens as the heel rotates off the floor pushing back the toes to start the toe-off phase. The first peak in shear force is considered to be the critical one with respect to slips resulting in falls. It occurs at about 19% into the stance phase, which is 90 to 150 ms after heel contact depending on stance duration.

The forces occurring shortly after heel contact have been thought to play an important role in slips and falls. Small spikes in the GRFs have been recorded (Perkins 1978, Lanshammar and Strandberg 1981, Whittle 1999) (figure 2). The first spike immediately after heel contact tends to be in the anterior or forward direction. This spike is not always evident, perhaps because of measurement error due to very low force levels or to low sampling rates. However, those who have recorded these forces have attributed them to the movement of the heel as it impacts the floor and transfers momentum to the ground (Perkins 1978, Whittle 1999). Another peak in the shear GRF that may be more important to slips and falls occurs a little later and is in the rearward direction (Perkins 1978, Lanshammar and Strandberg 1981, Strandberg 1983). A more careful examination of the heel kinematics by

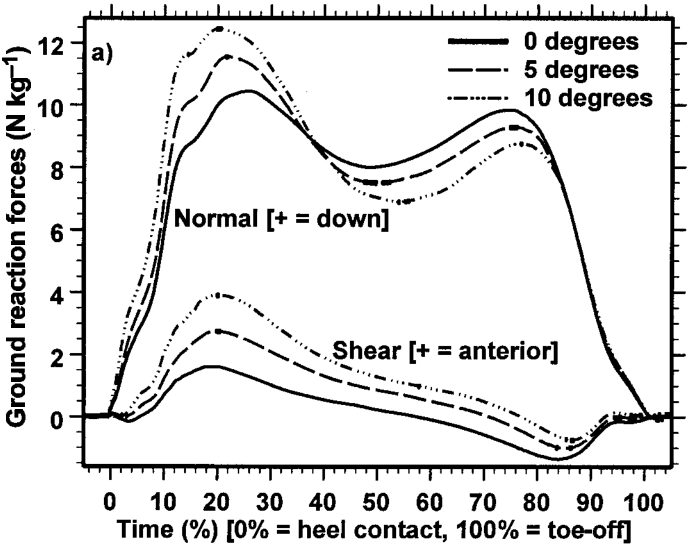

Figure 1. Shear and normal forces for walking along horizontal surfaces and down inclined surfaces (from Cham and Redfern 2001a).

Lanshammar and Strandberg suggested that this spike is due to rearward movement of the heel during the early loading phase. As the foot rotates down on the floor and reaches foot-flat position (at about 15% into stance), it creates another broader spike especially evident in the normal GRF (Whittle 1999).

Since the shear forces are highest near the heel contact and push-off phase (Redfern and DiPasquale 1997), these are the points where slips most often occur. Heel contact is the critical phase where slips can result in falls (Strandberg 1983, Rhoades and Miller 1988, Lloyd and Stevenson 1992, Redfern and Bloswick 1995, Hanson *et al.* 1999). Thus, the forces occurring at heel contact are of critical importance in determining if the frictional capabilities of the shoe/floor interface will be sufficient to prevent slips.

One GRF measure that has been used to quantify and understand the biomechanics of slips has been the ratio of shear to normal GRF components. During normal locomotion on dry surfaces, i.e. no-slip conditions, this ratio has been described as the 'required coefficient of friction' (RCOF) (Redfern and Andres 1984, Rhoades and Miller 1988, Grönqvist *et al.* 2001a). As a result of the normal and shear force profiles described earlier, the RCOF has a peak value occurring at about the same time as the peak shear force. This peak value is about 0.20 (McVay and Redfern 1994). The peak RCOF has been suggested to predict slip potentials for various gait activities (Redfern and Andres 1984, Love and Bloswick 1988, Buczek *et al.* 1990, McVay and Redfern 1994, Buczek and Banks 1996).

2.1.2. *Kinematics of walking*: Walking speed is an obvious characteristic that will impact slip potential. Laboratory measurements of self-chosen gait speeds have ranged from 0.97 m s^{-1}(Redfern and DiPasquale 1997) to 1.51 m s^{-1} (Murray *et al.* 1967) for level surfaces. Sun *et al.* (1996) reported walking speeds of 1.1 – 1.2 m s^{-1} on a considerably large number of subjects in a natural urban setting. Step length

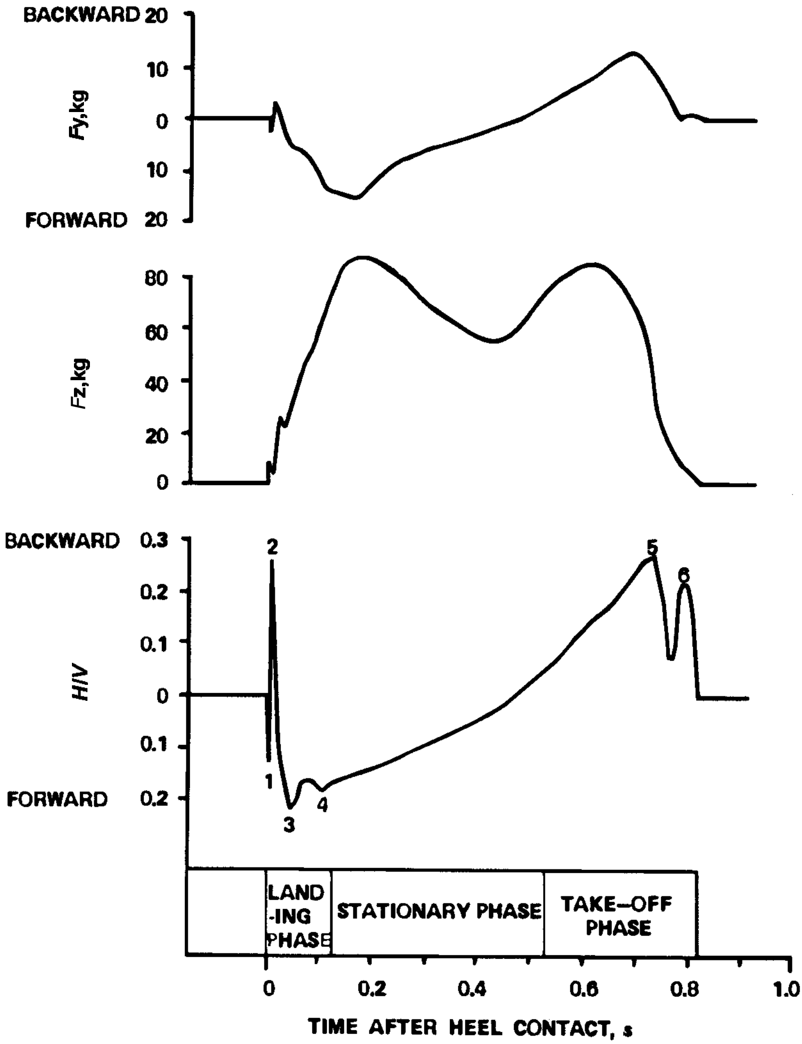

Figure 2. Typical ground reaction forces (F_y = anterior-posterior shear and F_z = normal) and required coefficient of friction (H/V) during stance. Note that peak 1 is caused by the forward force of impact of the heel onto the force plate. Peak 2 is a result of a backward force exerted on the heel after contact during the early landing phase. Peaks 3 and 4, often recorded as one broad spike, are caused by the main forward force, which retards the motion of the foot. Finally, peaks 5 and 6 are recorded during the push-off phase, with the toes in contact with the force plate, pushing in the backward direction (from Perkins 1978).

theoretically has an important effect on slip potential. The influence of step length on slip potential was explored using a static model by Grieve (1983). As the step length is increased, the ratio of shear to normal forces at heel contact would change, resulting in a greater shear force during the initial portion of the step. Thus, reducing step length is one method that can reduce the slip potential when walking.

The kinematics of the heel as it comes in contact with the floor is believed to have a role in the potential for slips and falls (Redfern and Bidanda 1994). Recordings of heel movements have shown that the heel rapidly decelerates just prior to heel contact, then there is a slight sliding motion along the surface at impact (Strandberg and Lanshammar 1981, Perkins and Wilson 1983, Cham and Redfern 2001a). (See solid line in figure 3a for level walking.) The patterns of sliding during this time can be variable. In general, studies have shown that the heel velocity is forward immediately upon impact, then either coming to a stop or reversing sliding direction before coming

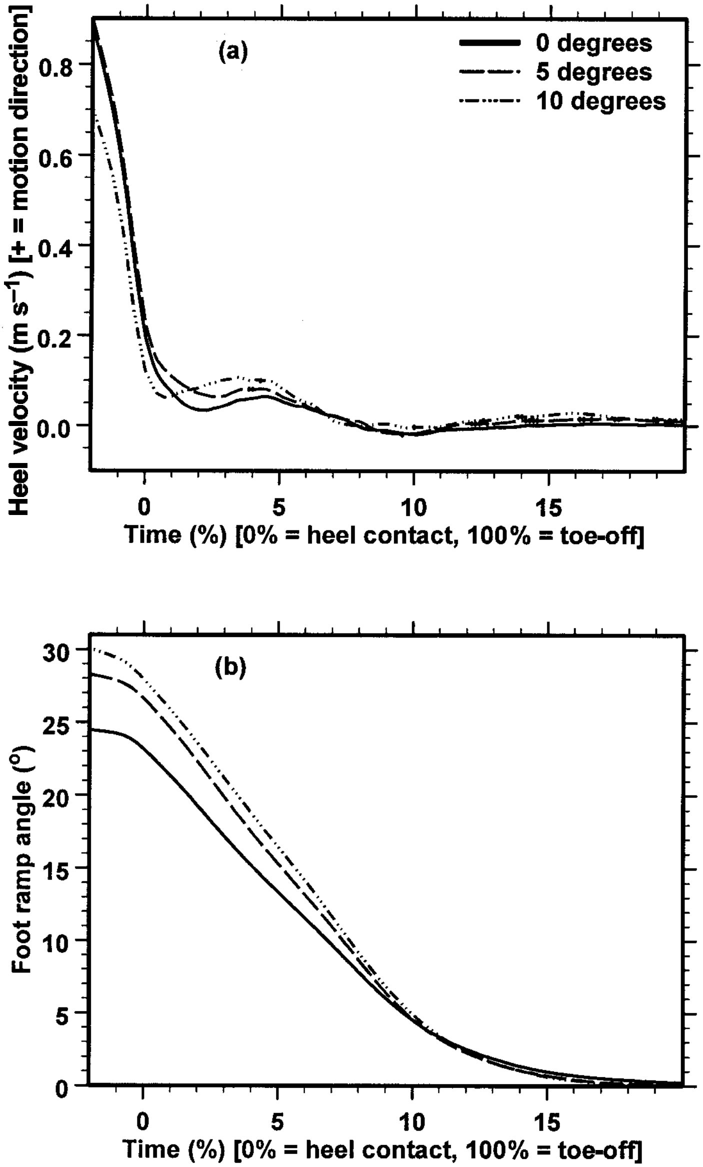

Figure 3. Characteristic profile of (a) the heel velocity, and (b) the foot-ramp angle (0°, 5°, 10°), averaged across trials conducted on dry vinyl tile flooring. (Time is truncated at 20% of the stance for a more detailed view of heel contact dynamics.) Adapted from Cham and Redfern 2001a.

to a stop. However, Cham and Redfern (2001a) reported a significant number of walking trials where the heel's impact velocity in the antero-posterior direction was negative (evident in the standard deviation associated with the heel velocity in table 2), i.e. the heel was moving in the rearward direction at the instant of contact. In all reported cases, this rapid heel motion ended shortly after heel contact and the heel came to a complete stop, while the foot continued to rotate down on the floor from about 23° for level walking, reaching a foot flat position about 15% into stance.

Joint angles were investigated by Murray *et al.* (1967), Winter (1991) (level walking), Redfern and DiPasquale (1997), and Cham and Redfern (2001a). In general, the overall profiles of joint angles were in agreement across studies. (See solid line in figure 4 for level walking angle profiles.) At heel contact, the ankle is in slight dorsiflexion but rapidly reaches its peak plantar flexion angle (around 10% of stance phase) as the foot rotates down onto the floor. Toe-off phase begins about 80% into stance when the heel comes off the floor and the ankle again goes into plantar flexion. During the first 30% of the stance, there is an increased flexion of the knee, caused mainly by the forward rotation of the shank. During the last phase of the stance ($>60-80\%$), characterized by the movement of the body's centre of gravity past the single leg base of support, the knee flexes again as the subject prepares for the heel contact of the other foot (second half of the non-supporting leg's swing phase) and toe-off of the supporting foot. The hip angle profile reflects the changes in the upper leg orientation, i.e. only small variations in torso orientation (a few degrees) have been recorded. For most of the stance duration the hip is in extension due to the continuous forward rotation of the upper leg. However, at the end of the stance phase, the subject prepares for the swing phase by rotating the foot off the floor, flexing the knee and the hip (via rearward rotation of the upper leg).

2.1.3. *Joint moments*: One biomechanical measure that sometimes has been overlooked with regards to slips and falls is joint moments. Moments at the ankle,

Table 2. Foot and heel kinematic parameters for normal walking on level dry surfaces (0°) and inclinded surfaces (5° and 10°).

Variable: Mean (SD)	0°	5°	10°
Heel velocity in the direction of motion at HC (m s^{-1})	0.19 (0.39)† 0.14 (0.27) – 0.68 (0.52)¶ 1.03 (0.16)†† during 60 ms prior to heel contact 0.3 – 2.75‡‡	0.25 (0.42)†	0.13 (0.32)†
Foot angular velocity at HC (° s^{-1})	223.8 (98.4)†	251.7 (111.9)†	292.9(86.9)†
Heel contact angle (°)	23.5 (3.7)† $\approx 30^\ddagger$ 10 – 30§ 22 (5.3)¶ 32 (4)†† during 60 ms prior to heel contact	26.4 (3.5)†	26.9 (4.9)†

From †Cham and Redfern (2001a) [vinyl floors], ‡Leamon and Son (1989), §Perkins (1978), ¶Strandberg (1983) ['grip' trials and heel velocity averaged within subjects], ††Gronqvist (1999), ‡‡Morach (1993).

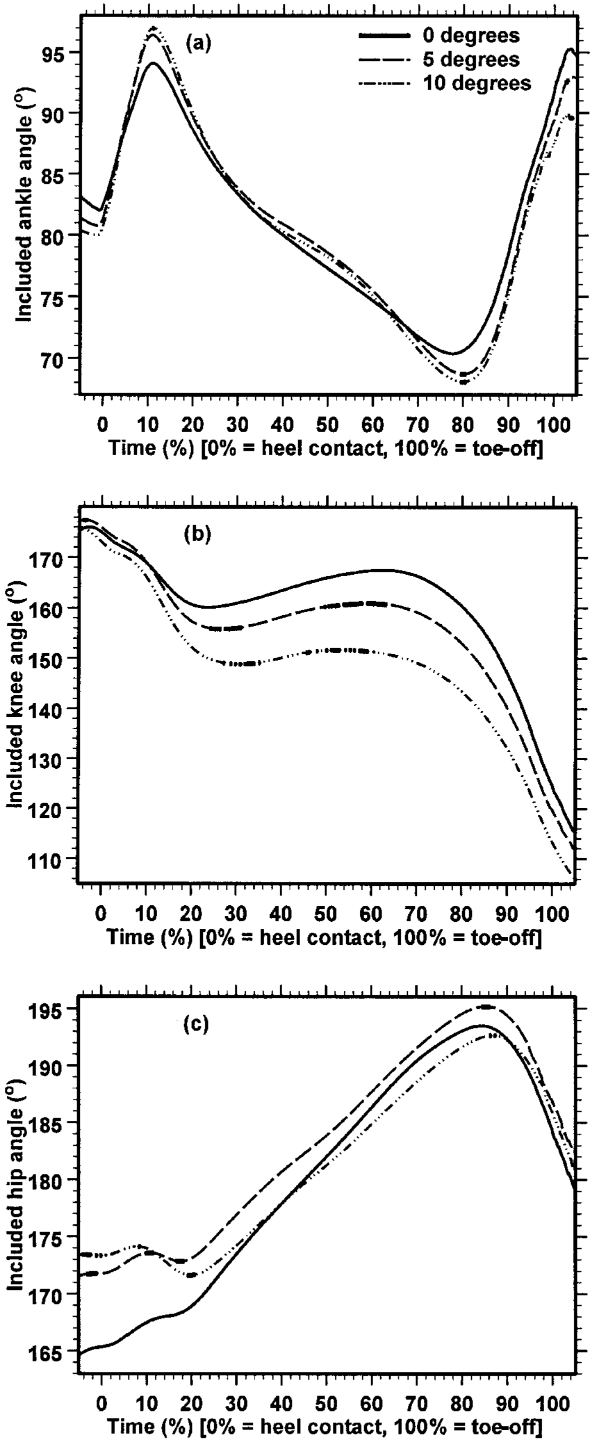

Figure 4. Characteristic profile of included joint angles during gait on the vinyl tile floor (0°, 5°, 10°), averaged across all trials: (a) ankle, (b) knee, and (c) hip (adapted from Cham and Redfern 2001a).

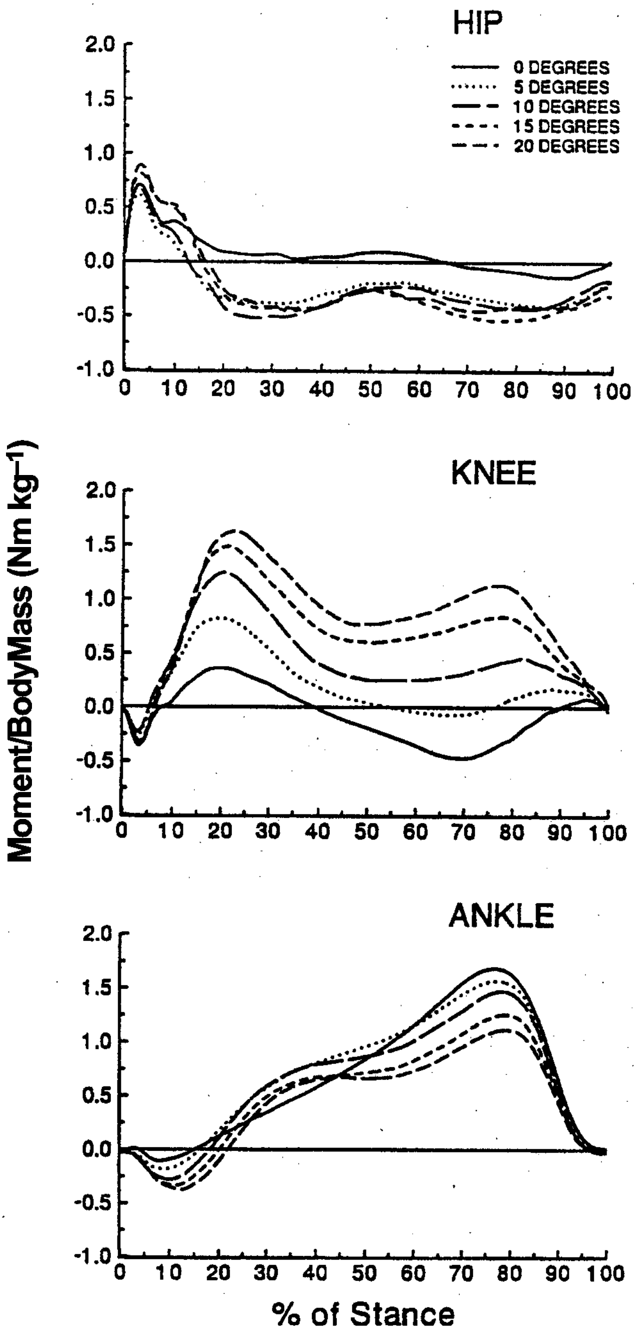

Figure 5. Characteristic profile of joint moments (normalized to body weight) during gait on level and inclined surfaces (from Redfern and DiPasquale 1997).

knee and hip are required to maintain upright walking. These moments are related to the strength level required to walk, and actions to recover from a slip if necessary.

The moments at the ankle, knee and hip joints have been calculated for level walking by a number of researchers for a variety of subject populations (Winter 1991). For a healthy population, the moments for level walking are shown in the solid line of figure 5. Note the biphasic (extension-flexion) nature of the knee moment and increasing plantar flexion moment at the ankle.

2.2. *Walking on an inclined surface*

Walking on an inclined surface, i.e. ramp, changes the characteristics of gait, and therefore the potential for slipping. Walking down a ramp increases the risk of slips and falls much more than walking up because of the increased shear forces generated and the reduced ability to recover should a slip occur. For example, Haslam and Bentley (1999) reported that falls in postal workers occurred 30% of the time walking down a sloped drive compared to 2% walking up. This section presents the changes that occur in the biomechanics of gait when walking down a sloped surface, which increases the risk of slips and falls.

2.2.1. *Ground reaction forces on inclined surfaces*: Changes in the inclination of the floor, i.e. increasing ramp angle, are associated with changes in the ground reaction forces (Redfern and DiPasquale 1997, Cham and Redfern 2001a) (figure 1). For example, shear forces for level walking reach a maximum of about 1.5 to 1.8 N kg^{-1} (normalized to body weight). However, walking down a ramp increases this peak shear by about 61% for a 5° ramp angle and 128% for a 10° ramp angle. The timing of these peak shear forces on ramps appears to be the same as walking on level surfaces. The normal forces also are affected by inclination angle, with an increase in the peak force of about 1 N kg^{-1} for a 5° increase (table 1). The peak normal force occurs earlier on inclined surfaces, leading to a time difference of about 5% between the peak of shear and normal foot forces on level surfaces, and an almost in-phase foot force when the ramp angle was increased to 10°. All these changes affect the RCOF, with the peak RCOF increasing with inclination of the surface. For example, walking down a 20° ramp creates an increase in the peak RCOF from the level value of 0.18 to 0.45 (figure 6). Table 1 shows the increase in the GRFs and peak RCOFs as ramp angle is changed. Ramp angle also has an effect on RCOF for walking up (McVay and Redfern 1994); however, the peak RCOF occurs towards the end of the push-off phase when slips do not usually result in falls.

2.2.2. *Kinematics on inclined surfaces*: Walking down an inclined surface has been found to have an effect on some kinematics variables, but not on others. For example, natural gait velocity was not found to be significantly different when walking down a ramp compared to walking on a level surface (Sun *et al.* 1996, Redfern and DiPasquale 1997). However, step length was reduced as ramp angle was increased. More specifically, Redfern and DiPasquale (1997) reported a step length of 0.54 m and 0.48 m for level walking and during descent of a 20° ramp, respectively, while Sun *et al.* (1996) found these values equal to about 0.62 and 0.57 m for nearly-horizontal (2°) surfaces and 9° ramps, respectively.

The kinematics of the foot upon heel impact during the descent of inclined surfaces are similar to those for level walking, especially the profile of the heel's linear velocity along the floor surface and the sliding patterns of the heel along with the slip-distance from heel contact. Other variables were slightly more affected by the walkway inclination including foot-floor angle and foot angular velocities recorded

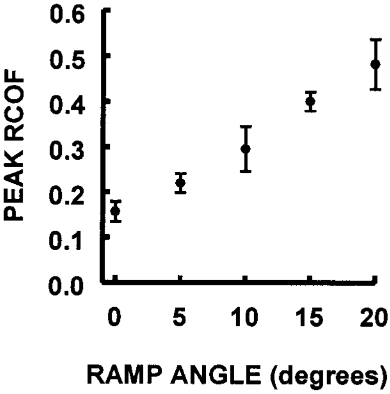

Figure 6. Peak required coefficient of friction (RCOF) as a function of ramp angle (adapted from Redfern and DiPasquale 1997).

at heel contact (table 2). The foot reaches foot-flat position at about the same time in the gait cycle (15% of stance) as when walking on a level surface. Surface inclination angle has an effect on joint angles during gait (figure 4). While there are minor changes at the hip and ankle, the included knee angle is most affected (Redfern and DiPasquale 1997, Cham and Redfern 2001a).

2.2.3. *Joint moments on inclined surfaces*: The inclination of a surface can have a significant effect on the moments at the lower extremity joints. Figure 5 shows typical moments during gait for walking on a level surface and the impact of ramp angle (Redfern and DiPasquale 1997). As depicted in figure 5, the resulting moments at ankle, knee and hip were found to change as a function of ramp angle, with the knee moment being the most affected by ramp angle. Redfern and DiPasquale (1997) reported an increase from a mean of 0.4 Nm kg^{-1} when walking on a horizontal surface to 1.7 Nm kg^{-1} when descending a 20° ramp.

2.3. *Walking on stairs*

From a functional standpoint, stair ambulation is a much more challenging task when compared to level gait or walking on ramps. While negotiating stairs, the body is carried in both a vertical and forward direction, which results in joint motion and muscular demands that differ significantly from walking. Walking on stairs not only challenges the strength and range of motion limits of the lower extremity, but also requires substantial balance and muscle co-ordination as well. Owing to the vertical nature of stair ambulation, a slip on stairs can result in a catastrophic event resulting in serious injury. A thorough understanding of the biomechanics during stair negotiation is important for understanding how slips and falls can be prevented during this high demand task.

2.3.1. *Ground reaction forces during stair ambulation*

2.3.1.1. *Ascending stairs*: As with level walking, the vertical GRF during stair ascent demonstrates two distinctive peaks, one during weight acceptance (i.e. 30% of stance) and the other during late stance (McFayden and Winter 1988). In contrast to level walking, however, the second peak tends to be slightly greater than the first (figure 7(a)). The higher second peak illustrates the increased force applied to the floor through strong contraction of the plantarflexors as the body is being elevated to the next step.

Ground reaction force patterns in the anterior-posterior and medial-lateral directions are similar to those during level walking. At weight acceptance, there is an anterior shear force acting on the floor while there is a posterior shear force acting on the floor at toe-off (McFayden and Winter 1988). As shown in figure 7(b), the anterior shear force tends to be somewhat greater in magnitude than the posterior shear force (10% vs 5% body-weight). There also is a lateral shear force acting on the floor that is fairly consistent throughout stance reaching a maximum value of approximately 5% body-weight (figure 7(c)).

As described earlier, the ratio of the resultant shear forces to the vertical (or normal) force has been described as the required coefficient of friction or RCOF. During stair ascent, the RCOF during weight acceptance has been observed to be consistent with values reported for level walking (0.21); however, the RCOF during toe-off appears to be somewhat higher (0.39) (figure 7(d)). Such data suggests that a slip would be more likely to occur in the posterior direction during late stance as the body is being elevated.

2.3.1.2. *Descending stairs*: The vertical ground reaction force pattern during stair descent varies significantly compared to that of stair ascent (McFayden and Winter 1988). The peak vertical ground reaction force (which also occurs during

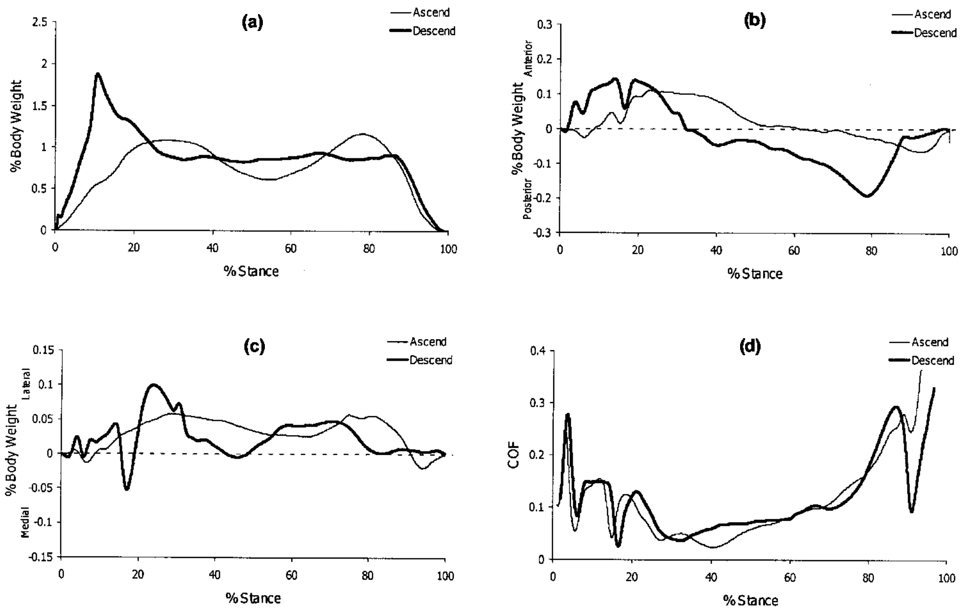

Figure 7. Representative ground reaction forces and required coefficient of friction data obtained during stair ascent and descent in a healthy adult subject (AMTI force plate, 2400 Hz): (a) vertical ground reaction force, (b) anterior-posterior ground reaction force, (c) medial-lateral ground reaction force, and (d) required coefficient of friction (resultant shear force/normal force). Coefficient of friction data below 50 N of vertical force omitted. Unpublished data, Musculoskeletal Biomechanics Research Laboratory, University of Southern California.

weight acceptance) is much higher than that of stair ascent (190% body-weight) (figure 7(a)). This greater value reflects the greater downward acceleration of the body as it is lowered to the next step. On the other hand, the second peak that occurs prior to toe-off is much lower than that of stair ascent (95% body-weight).

The anterior-posterior shear forces demonstrate the same biphasic pattern evident in level walking and ascending stairs; however, the absolute values tend to be somewhat greater (figure 7(b)). The peak anterior shear force acting on the floor during weight acceptance is approximately 15% body-weight, which is similar to the peak posterior force acting on the floor prior to toe-off. The lateral shear force acting on the floor is present also during stair descent with peak values being somewhat greater than those observed during stair ascent (approximately 10% of body-weight) (figure 7(c)).

Despite the fairly large differences in ground reaction forces during stair ascent and descent, the RCOF is quite similar. For example, the RCOF during weight acceptance remains approximately 0.26, while the RCOF just prior to toe-off reaches a maximum of approximately 0.34 (figure 7(d)). Although these data suggest that a slip also is more likely to occur just prior to toe-off, there does not appear to be any greater risk associated with descending stairs as compared to ascending stairs.

2.3.2. *Kinematics of stair ambulation*

2.3.2.1. *Ascending stairs*: During stair ascent, the greatest differences in joint motion (when compared to level walking) occurs at the knee and hip. As the foot makes contact with the stair, the hip and knee are flexed to approximately 60° and the ankle is in about 10° of dorsiflexion (McFayden and Winter 1988, Powers *et al.* 1997) (figure 8(b), (c)). Elevation of the body is accomplished through hip and knee extension, which peaks during late stance (50% of the gait cycle). Early stance-phase ankle dorsiflexion permits tibial progression and accommodates the increased requirement of knee flexion (figure 8(a)). Increased flexion of the hip (60°) and knee (8°) are required during swing to clear the foot, and to place the limb in the appropriate position in preparation for contact with the next step (McFayden and Winter 1988, Powers *et al.* 1997). Swing phase motion of the ankle is similar to that of level walking.

2.3.2.2. *Descending stairs*: During stair descent, contact with the lower step is made with the ankle in approximately 20° of plantarflexion and the knee and hip slightly flexed (10° and 20°, respectively) (Powers *et al.* 1997) (figure 8). Lowering of the body is primarily accomplished through knee flexion, which peaks at about 80° by the end of stance (McFayden and Winter 1988, Powers *et al.* 1997) (figure 8(b)). Hip flexion (30°) also contributes to lowering of the body (figure 8(c)). Progressive ankle dorsifexion is evident throughout stance, reaching a maximum of 15° by approximately 50% of the gait cycle (figure 8(a)). In anticipation for contact with the next step, progressive hip and knee extension is evident and the ankle plantarflexes during swing (Powers *et al.* 1997).

2.3.3. *Joint moments during stair ambulation*

2.3.3.1. *Ascending stairs*: The muscle moments generated at the lower extremity joints during stair ambulation vary significantly compared to level walking. Elevation of the body is accomplished through large hip and knee extensor

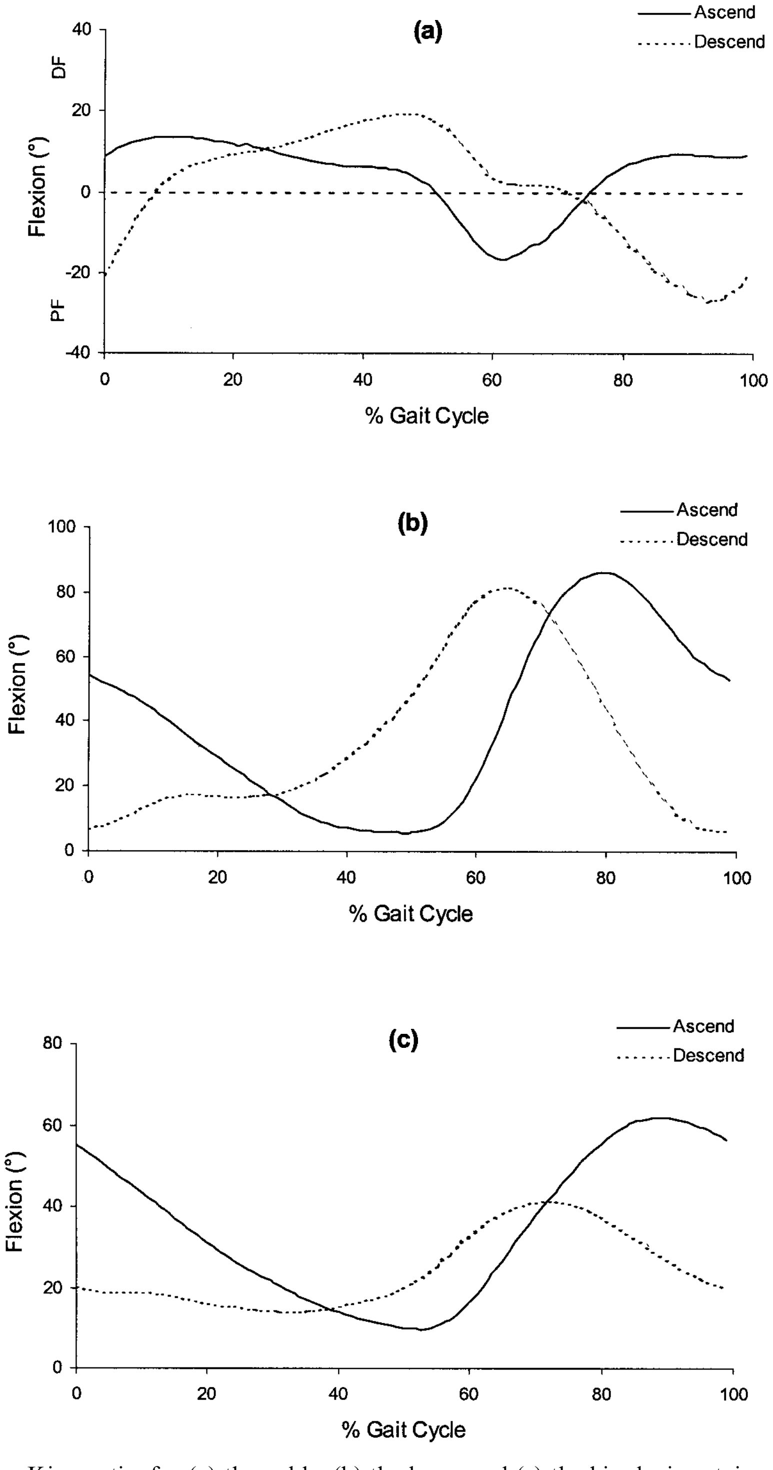

Figure 8. Kinematics for (a) the ankle, (b) the knee, and (c) the hip during stair ascent and descent. Ensemble averaged curves obtained from healthy adults (n = 19). Data taken from Powers *et al.* (1997). DF = dorsiflexion, PF = plantarflexion.

moments, which peak during early stance (approximately 1.0 Nm kg^{-1} at both joints) (Andriacchi *et al.* 1980, McFayden and Winter 1988, Salsich *et al.* 2001) (figure 9(b), (c)). During stair ascent, the ankle demonstrates a plantarflexor moment throughout stance peaking just prior to swing (1.5 Nm kg^{-1}) (Andriacchi *et al.* 1980, McFayden and Winter 1988) (figure 9(a)). The plantarflexor moment controls tibial rotation and provides for push-off during stance.

2.3.3.2. *Descending stairs*: During stair descent, the lowering of body-weight is accomplished primarily through a knee's strong extensor moment (Andriacchi *et al.* 1980, McFayden and Winter 1988, Salsich *et al. in press*). The knee moment demonstrates two peaks, one during weight acceptance (early stance) and a larger extensor moment (1.0 Nm kg^{-1}) as the body is being lowered in late stance. The ankle contributes significantly to stability during stair descent as evidenced by a large plantar flexor moment that peaks early during stance (1.4 Nm kg^{-1}) (Andriacchi *et al.* 1980, McFayden and Winter 1988). The hip demonstrates a relatively small extensor moment throughout stance peaking at 0.2 Nm kg^{-1} during weight acceptance (Andriacchi *et al.* 1980, McFayden and Winter 1988).

2.4. *Load carrying using the standard industrial symmetrical 2-handed posture*

Many workers are required to carry loads as part of their daily occupational tasks. Over the years, load lifting and holding tasks have been the focus of research directed towards preventing load handling-related musculoskeletal injuries. For load carrying, researchers have concentrated mostly on muscle activity patterns and physiological strain parameters and less on gait biomechanics and stability parameters. Cham and Redfern (2001b) investigated the effect of carrying relatively light loads (no load, 2.3 and 6.8 kg) on slips- and falls-related gait biomechanics during normal locomotion. In this investigation, load carrying was associated with small but significant decreases in the required frictional properties for safe walking, a finding that was previously reported by Love and Bloswick (1988) for level walking. Kinematic changes associated with load carrying reported by Cham and Redfern (2001b) included minor postural adaptations such as increased knee flexion and slower heel contact velocity along the floor surface.

Myung and Smith (1997) have examined load-carrying effects on specific parameters such as step length and heel velocity during level walking. The authors reported a significant decrease of stride length with increasing load levels, a result that was not confirmed by Cham and Redfern (2001b). This apparent disagreement could be due to the different load levels considered in the two studies. In Myung and Smith (1997) the load level ranged from the no-load condition to 40% of body weight, a far greater load level than the ones investigated by Cham and Redfern (2001b). Myung and Smith (1997) have also concluded that heel velocity at heel contact was not affected by load carrying levels on dry floors, a finding that apparently contradicts Cham and Redfern's results. Differences in the two experimental and analysis procedures could be responsible for this apparent contradiction in the results: (1) Myung and Smith (1997) had a fixed walking speed while Cham and Redfern (2001b) had a natural pace, and (2) Myung and Smith (1997) used the resultant vector of heel velocity, while Cham and Redfern (2001b) investigated the individual components (anteroposterior and lateral) along the floor surface.

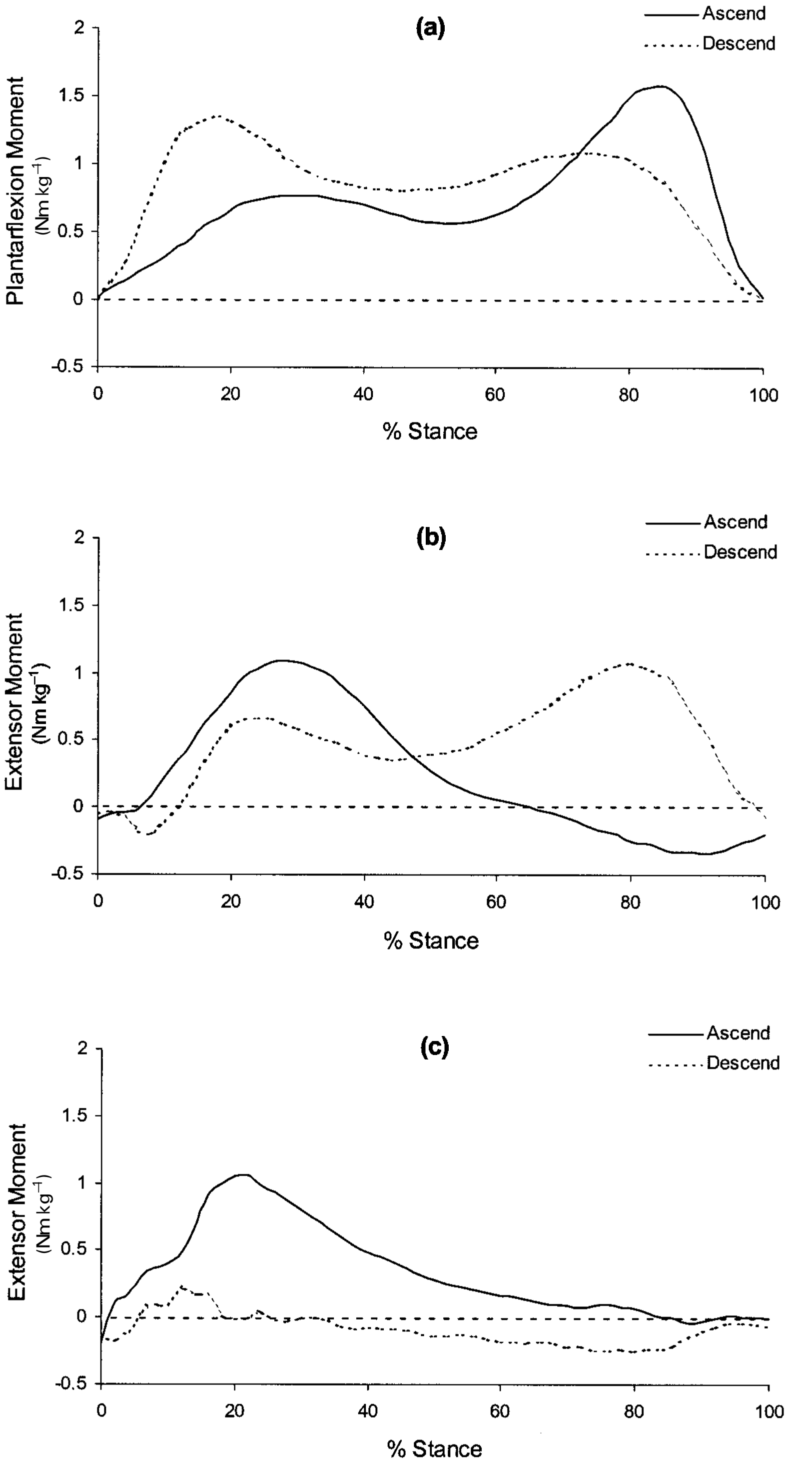

Figure 9. Net joint moments for (a) the ankle, (b) the knee, and (c) the hip during stair ascent and descent. Ensemble averaged curves obtained from healthy adults ($n = 10$). Data taken from Salsich *et al.* (2000).

3. Biomechanics during slipping

3.1. *Definition of slip from a biomechanical perspective: microslips/macroslips and falls*

During normal gait on dry, non-slippery surfaces, heel sliding along the floor surface has been observed at and shortly after heel contact before coming quickly to a complete stop. This heel motion characterized as 'normal' (Perkins 1978, Cham and Redfern 2001c) or termed 'grip' (Strandberg and Lanshammar 1981) or 'microslip' (Perkins 1978, Leamon and Son 1989), is not detected by subjects. Based on the distribution of slip distances on dry surfaces or the human perception of slipping (Leamon and Li 1990), researchers have used cut-off values of 1 cm (Perkins 1978, Cham and Redfern 2001c) or 3 cm (Leamon and Li 1990) above which the outcome of a contaminated trial was classified as a full slip or 'macroslip' (table 3). Standberg and Lanshammar (1981) used a somewhat more detailed categorization of contaminated surface trials that did not develop into falls (slip-stick). The so-called slip-stick trials were further divided into three groups: mini-slip (subjects did not detect the slipping motion), midi-slip (slip-recovery trials without 'major gait disturbances') and maxi-slip (slip-recovery with large corrective responses or 'near-fall' trials). As expected, slip distance and peak forward sliding velocity increase with the severity of the slip (table 3). Redfern and colleagues (Hanson *et al*. 1999, Cham and Redfern 2001c) have categorized trials as falls in two cases: (1) the heel kinematic data showed that the heel did not come to a stop after heel contact, and/or (2) the subject lost balance and eventually fell into the safety harness.

3.2. *Kinematics of walking on slippery surfaces*

Biomechanical human reactions to slippery surfaces partially determine the outcome of slipping perturbations (no slip, slip-recovery, slip-fall), and are therefore important to monitor for understanding the complex relationship between gait biomechanics and actual slips and falls incidence. The descriptions of typical slip-recovery events appear consistent across studies (Perkins 1978, Strandberg 1983, Cham and Redfern 2001c), although the magnitude and timing of gait parameters are not always given. Typically, trials leading to a slip event are characterized by higher linear impact heel velocities (not always consistent across Strandberg's subjects), slower foot angular velocities at heel contact and faster sliding heel movements after heel contact, when compared to dry or grip trials (table 3). Generally, subjects are able to slow down the heel to very low velocity levels, often even sliding in the rearward direction. Cham and Redfern (2001c) reported that subjects were always able to rotate their foot down onto the floor and reach foot-flat position regardless of the trial's outcome. As pointed out by Perkins (1978), Strandberg (1983), and Cham and Redfern (2001c), the forward slip starts slightly after heel contact (about 50–100 ms) (for example, see figures 10 and 11). Strandberg (1983) suggested that a slip is likely to result in a fall if the slip distance is in excess of 10 cm or the peak sliding velocity is higher than 0.5 m s^{-1}. Cham and Redfern (2001c) reported heel velocity of slip-fall outcomes reaching a local maximum (table 3) before subjects attempted to control the slipping motion thus slowing the heel's sliding motion, sometimes even reversing it (as shown in figure 11) to a local minimum of heel. At that time, the heel accelerates again and eventually leads to a fall. This attempt to recover has not been reported by Perkins (1978), but is evident in data presented by Strandberg (1983), although not discussed.

Table 3. Definition and characteristics of trial outcomes for level walking.

Slip distance (SlipDist) (cm) Peak forward velocity of slip (MaxVel) ($m\ s^{-1}$)	Perkins†	Strandberg‡		Leamon and Li§	Grönqvist¶	Cham and Redfern††	
	SlipDist (cm)	SlipDist (cm)	MaxVel (m/s)	SlipDist (cm)	SlipDist (cm)	SlipDist (cm)	MaxVel (m/s)
Normal-slip, microslip or grips on dry surfaces	< 1.0	NA	NA	< 3.0	NA	< 1.0	0.10 (SE 0.01)
Macroslip, slip, slip-recovery,slip-stick	Few cm	Mini-slip 1.2 (0.4) Midi-slip 5.1 (4.7) Maxi-slip > 8.6 (3.7)	0.23 (0.04) 0.49 (0.22) 0.56 (0.50)		1.6 (0.4) during 60 ms after heel contact 2.7 (4.7) 60 – 200 ms after heel contact*	4.0 (SE 3.0)	0.31 (SE 0.06)
Fall, skid-fall	> 10 – 15	NA	> gait speed (1 – 2 $m\ s^{-1}$)		NA	No recovery when SlipDist > 10.0	0.78 (SE 0.16)**

†Perkins (1978), ‡Strandberg (1983), §Leamon and Li (1990), ¶Grönqvist (1999), ††Cham snd Redfern (2001c).

*One slip (out of 12 trials) was a likely fall since the total slip distance was 19 cm and the peak slip velocity was 2.1 $m\ s^{-1}$ (160 ms after heel contact). All other slip distances were less than 6 cm (range 1.5 to 5.9 cm) and peak slip velocities were less than 0.55 $m\ s^{-1}$ (range 0.22 to 0.54 $m\ s^{-1}$).

**For fall trials, MaxVel derived before subject attempts a recovery, at which time the heel velocity is brought to a minimum before increasing again, leading to a fall.

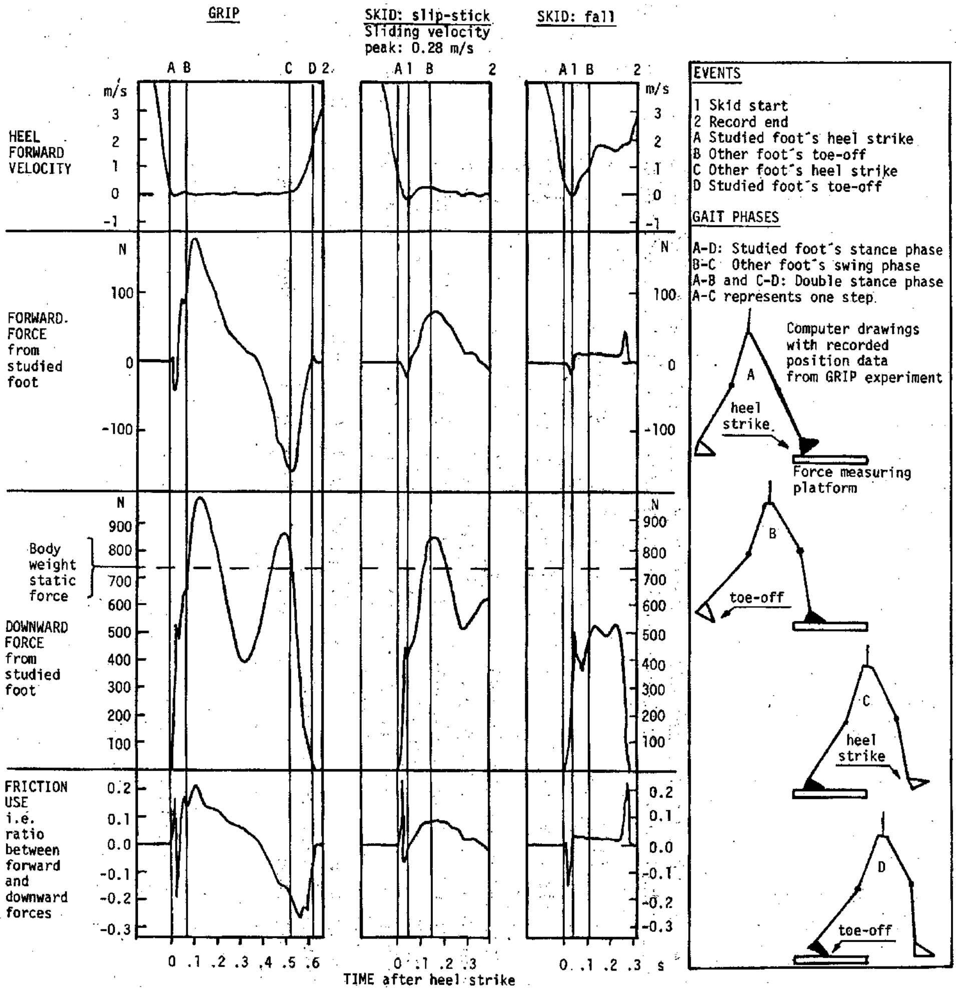

Figure 10. Typical characteristic profiles of heel velocity and ground reaction forces in the direction of motion reported by Strandberg (1983).

3.3. *Ground reaction forces during slips*

GRF profiles appear to be more varied than kinematic variables across slip trials. However, general characteristics can be identified. First, both peak shear and normal GRFs are reduced during slip events (Strandberg 1983). Second, the transfer of body weight to the supporting leg does not seem to be completed in fall trials. This is evident not only in the shape of the normal forces (Strandberg 1983), but also in the progression of the centre of pressure, which stayed close to the ankle in fall cases (Cham and Redfern 2001d). Third, after a slip has developed, a corrective response or attempt at bringing the foot back near the body, can sometimes be identified as associated with a decrease in the shear forces (25–45% into stance) (Cham and Redfern 2001d).

As mentioned in section 2.1.1, on dry surfaces, the shear to normal force ratio or RCOF has been interpreted as the frictional requirements for a no-slip outcome

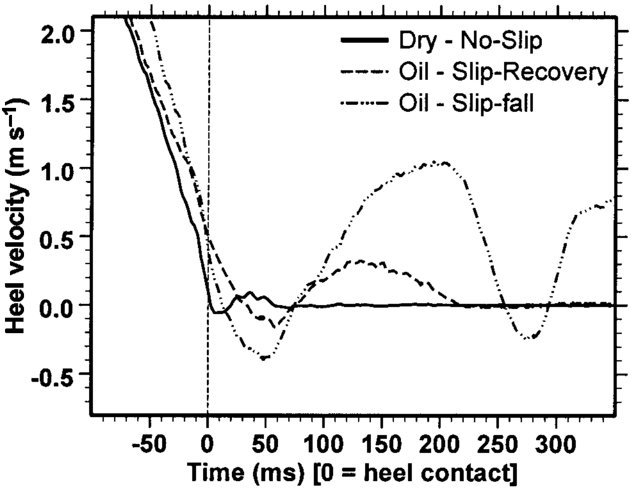

Figure 11. Typical examples of linear heel velocity profile recorded along the direction of motion during dry (no-slip) and oily (slip-recovery and slip-fall) conditions (from Cham and Redfern 2001c).

(Perkins 1978, Strandberg 1983, Redfern and DiPasquale 1997). During slippery trials, the shear to normal ratio was defined as the instantaneous utilized friction (Strandberg 1983) or achievable friction (Hanson *et al.* 1999), which decreased with the severity of the slip (Strandberg 1983, Grönqvist *et al.* 1993). For soapy conditions and level surfaces, for example, Strandberg (1983) has reported achievable COF (ACOFs) as low as 0.02 (a fall case) up to 0.15 (mini-stick), compared to the peak RCOF value of 0.17 for grip trials. Grönqvist found that the time-averaged (100–150 ms after heel contact) ACOF during slippery trials (for seven male subjects) on a level surface was 0.11 ($\pm$0.04) for the 'slip-recovery' condition and 0.04 ($\pm$0.02) for the 'slip-fall' condition (Grönqvist *et al.* 1993).

4. Postural control

4.1. *Joint moment response during slips*

Dynamic analyses have been used in many biomechanical studies on a number of activities such as lifting or carrying to determine the net moments at various joints. However, the joint moments of force of the lower extremity have been most widely researched in gait and balance studies. Moment patterns vary at the hip and knee during walking as a result of the balance control of total limb synergy (Winter 1995). The large changes in hip and knee moment patterns seen during gait on level and inclined surfaces (figure 5), serve not only to generate the power necessary for the task, but also must keep a proper inter-segmental relationship to maintain balance. If the moments are not distributed and co-ordinated among the joints properly, balance would be compromised. Joint moments generated during a slip reflect an attempt of the person to bring the body back into equilibrium.

Joint moments in response to slipping represent the biomechanical reactions to maintain or recover balance. A steady gait pattern will be interrupted at the onset of slip and a rapid balance recovery attempted. This recovery attempt, sometimes termed a protective stepping strategy, often included large moment deviations from the steady gait pattern since the occurrence of such stepping is often unexpected and unrehearsed. Cham and Redfern (2001d) have investigated corrective strategies adopted in an attempt to avoid a fall after a slipping perturbation on oily level surfaces. Increased flexion moment at the knee was identified as the dominant response to slips between 25 and 45% into stance (figure 12(b)). Coincidently, the moment generated at the hip reflected a bias towards extensor activity (figure 12(c)). The ankle joint, on the other hand, acted as a passive joint during fall trials (figure 12(a)). This is due to the centre of pressure's proximity to the heel throughout stance in the fall cases, indicating, as mentioned previously, an uncompleted body weight transfer to the leading foot. Cham and Redfern (2001d) also reported that the corrective movements produced by those moments included increased knee flexion reaction, allowing subjects to rotate the shank forward, restore the ankle angle profile in an attempt to bring the foot back near the body, an effect that was evident in the deceleration of the sliding heel even in the case of slips resulting in falls (figure 11).

4.2. *Postural strategies*

Even though upright posture is inherently unstable and once a slip occurs a fall may appear to be unavoidable even among young adults (Pai 1999), humans still have a wide range of biomechanical responses available to protect themselves from actually falling to the ground. These responses can be both volitional involving conscious efforts and/or automatic involving reflexive reactions. They can be proactive as well as reactive movement strategies that can be implemented prior to, during, or after the loss of balance are experienced (Patla 1993, Woollacott and Tang 1997). Proactive control mechanisms are those that take place before the body encounters a potential threat to stability. A good example is the early detection and avoidance of potentially hazardous situations prior to actual contact (Woollacott and Tang 1997, Tang *et al.* 1999). Another example is the significant reduction in the peak RCOF recorded by Cham and Redfern (2001a) during trials when subjects anticipated the possibility of slips. The RCOF under these conditions were reduced by 16–33%. In addition, the perception of the danger of slipping affected the loading rate on the supporting foot, the joint moments and foot-floor angle at heel contact (Cham and Redfern 2001a).

Most slip and fall incidences occur unexpectedly. After the onset of a slip, a wide range of protective responses can involve both upper and lower extremities, such as grasping, arm swing, hip and ankle motion (ankle/hip strategy) (Horak and Nashner 1986, Horak 1992), and compensatory stepping (Maki and McIlroy 1997), as well as trunk motion. Grasping can be a quite effective recovery response, but one obvious limitation of the grasping strategy is the potential lack of any 'graspable' fixtures where the fall occurs. Even though the correction generated by ankle/hip movement can be produced in standing (Gielo-Perczak *et al.* 1999) or similarly during gait (Woollacott *et al.* 1999), it is often insufficient for protection against a fall. Larger disturbances in standing balance can seldom be restored without the subject's taking a step (Maki and McIlroy 1997). Thus the stepping response has a unique and irreplaceable importance in fall prevention.

The ability to recover is most likely determined by multiple factors in an interactive relationship. It is unclear, however, what factors determine the success rate of recovery in the protective stepping response after onset of a slip, what are the 'tradeoffs' between these factors, and furthermore what are the threshold values (or a

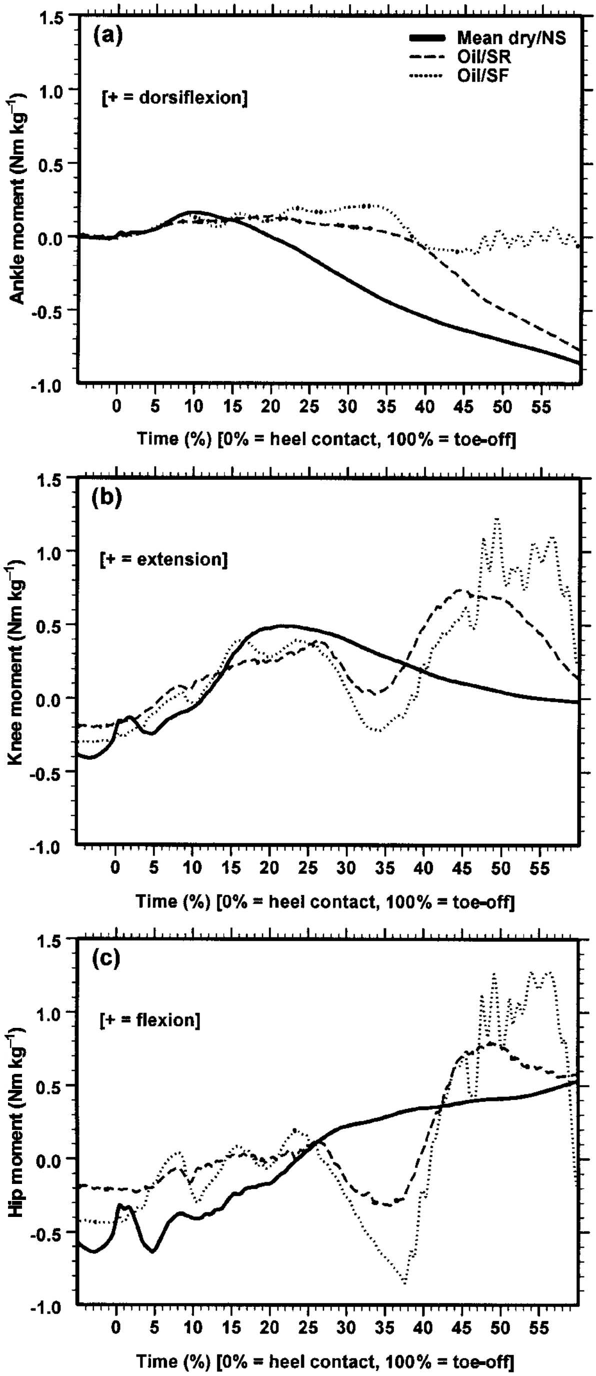

Figure 12. Mean profile of muscle moments generated at the lower extremity joints during stance phase on dry floors compared to typical profiles recorded during slip-recovery (SR) and slip-fall (SF) events on oily floors: (a) ankle moment, (b) knee moment, and (c) hip moment. The ankle moment decreased with the severity of the slip. Knee flexor and hip extensor moments were responsible for corrective reactions attempted between 25 and 45% into stance during slip events (from Cham and Redfern 2001d).

range of threshold values) beyond which a fall is mostly unrecoverable. These factors will probably include those affecting the relationship between the COM and base-of-support (BOS), such as the motion state (distance and velocity) of the slipping foot (Strandberg and Lanshammar 1981, Brady *et al.* 2000, Pavol *et al.* 2000) (figure 13) as well as the step length of the recovery limb (Hsiao and Robinovitch 1999).

4.3. *Joint stiffness control*

Studies of the dynamic behaviour of responses to perturbation while standing have been used to understand the complex responses of persons to slips while walking. Numerous studies by Winter (Winter 1990, Winter *et al.* 1998) have suggested that a body behaves like an inverted pendulum during the initiation of gait or perturbation and the centre of mass (COM) of the body is regulated through movement of the centre of pressure under the feet. They have postulated that the centre of pressure is controlled by ankle plantarflexor/dorsiflexor moment in the sagittal plane and hip abductor/adductor moment in the frontal plane. Winter *et al.* (1998) proposed a relatively simple control scheme for the regulation of upright posture that provides an almost instantaneous corrective response and reduces the operating demands on

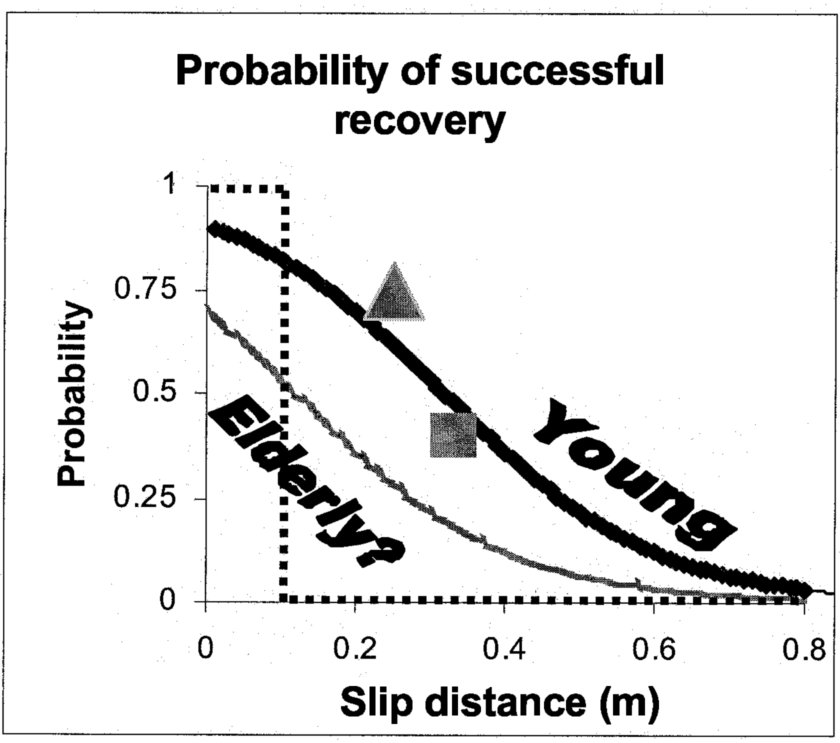

Figure 13. Strandberg and Lanshammar (1981) suggested a slipping distance of 0.1 m to be the likely threshold for a fall (dotted line). Recent work has shown much higher threshold values (thick line based on Brady *et al.* 2000), where recovery rate reduces from approximately 75% at 0.2 m to just over 10% at 0.6 m. These results are very similar to the observation made elsewhere (square and triangle, Pavol *et al.* 2000). The differences may result from the discrepancies in the methodology. It may be further hypothesized that the recovery rate will be further reduced among the older adults (thin line).

the central nervous system (CNS). Using these assumptions, Gielo-Perczak *et al.* (1999) proposed a mechanical structure of body response, which illustrated the combined effects of stiffness and damping of a subject on the strategy of the control of upright posture. Results showed that three types of postural strategies were performed during quiet standing in a frontal plane at: (1) the ankle joint; (2) the hip joint; (3) a combined strategy using both the hip and ankle joint together. In addition, these postural responses were found to depend on the type of perturbation and joint stiffness. It was observed that the nature of the perturbation must be known by the nervous system before joint stiffness was established. By adjusting joint stiffness, the resonant frequency is shifted to reduce potential resonance from the perturbation. Gielo-Perczak *et al.* (1999) concluded that joint stiffness control is used to help to maintain balance in response to slipping perturbations during walking on slippery surfaces.

5. Conclusions and future research

The biomechanics of slips and falls are an important component in the prevention of injury. This information can be used to develop slip resistance testing methodologies to reflect the frictional properties actually encountered in locomotion. In addition, biomechanical investigations of slips and falls can isolate specific events and times that are critical in determining the differences between slips leading to recovery and those leading to falls. One of the most critical biomechanical factors in slips and falls is thought to be the development of foot forces as the foot comes in contact with the ground. These forces (particularly the shear forces) must be counteracted by the properties of the shoe/floor interface. The ratio of the shear to normal foot forces generated during gait (known as the 'friction used' or RCOF) has been one biomechanical variable most closely associated with the measured frictional properties of the shoe/floor interface (usually the coefficient of friction, COF). Comparing this aspect of gait with measured shoe-floor properties appears to hold great promise for understanding the relationship between walking and potentials for slips and falls. However, there are other biomechanical factors in walking and slipping that also play an important role, such as the kinematics of the foot at heel contact. Slips of the heel naturally occur during most steps, with slip lengths of less than 1.0 cm. These slips (termed micro-slips) occur without the knowledge of the walker. This slipping action of the heel becomes correlated with actual slips noticed by the walker and falls as the slipping distance is increased. Linear motion of the foot coupled with rapid rotations at the ankle at about the time of heel contact make the actual dynamics and trajectories of the heel during these slips complicated, with motions occurring in the forward and rearward directions. The motions and forces at the foot are also variable, depending on the mental set of the walker. If there is a perceived danger of slipping, foot forces and kinematics will change (even if subjects are instructed not to do so). Thus, the biomechanics of walking are subject to the perceptions of the environment by the individual (Grönqvist *et al.* 2001b).

Future research on the biomechanics of slips will be needed to assist in reducing slip and fall injuries. Clearly, one area of future research is to expand our understanding of the shoe/floor contact interactions during slipping (Chang *et al.* 2001a, b). These data can then be used to develop a more 'biofidelic' slip resistance testing device that can measure friction under biomechanically relevant conditions. It is believed that testing slip resistance at the velocities, force levels, pressures and contact times seen in pedestrian walking will greatly increase the predictability of slips by these devices. Thus, knowledge of the kinetics of the foot during walking and slipping is necessary. A second related area of research is to conduct experiments where actual slips occur and relate the biomechanics to slip resistance measurements. While there have been a few studies that have measured the biomechanics of slipping, more research needs to be done. These studies have to carefully control the environments, the instructional set given to the subjects and the gait speed. In addition, data should be transferable to other experiments which are trying to assess the predictability of slip resistance testing devices.

Another future research direction is the investigation of human postural control strategies to prevent falls, including balance reactions when the environment is unknown and when it is known. This might include laboratory experiments investigating stepping responses or moment generations after a slip for different populations, including young adults, older adults, or persons with disabilities. Comparing the capabilities across populations will provide an understanding of the capabilities of people within different environments. The inclusion of research using computer model simulations in concert with the experimental studies may prove to be beneficial as well. The simulations under various conditions can reveal the biomechanical characteristics of the falls that have not been thoroughly demonstrated or understood. For example, one cannot *selectively* alter a subject's muscle strength or functional BOS in order to study the impact of reduced strength or functional BOS on movement stability. Nevertheless, such investigation can be readily performed with the aid of biomechanical model simulation.

A final area of suggested biomechanical research is the investigation of the mechanisms of falls and recovery to guide development of patient-based intervention strategies to prevent fall-related injuries. Such intervention can be achieved by improving both proactive and reactive motor responses. For example, biomechanical studies can be used to suggest methods for improving rehabilitation techniques for the elderly who are at risk for falls.

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