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The impact of a systematic reduction in shoe–floor friction on heel contact walking kinematics—A gait simulation approach

A. Mahboobin [a,*], R. Cham [a], S.J. Piazza [b]

[a] Department of Bioengineering, University of Pittsburgh, USA
[b] Departments of Kinesiology, Mechanical Engineering, and Orthopaedics & Rehabilitation, The Pennsylvania State University, USA

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ABSTRACT

Falls initiated by slips and trips are a serious health hazard to older adults. Experimental studies have provided important descriptions of postural responses to slipping, but it is difficult to determine why some slips result in falls from experiments alone. Computational modeling and simulation techniques can complement experimental approaches by identifying causes of failed recovery attempts. The purpose of this study was to develop a method to determine the impact of a systematic reduction in the foot–floor friction coefficient (μ) on the kinematics of walking shortly after heel contact ($\sim$200 ms). A walking model that included foot–floor interactions was utilized to find the set of moments that best tracked the joint angles and measured ground reaction forces obtained from a non-slipping (dry) trial. A "passive" slip was simulated by driving the model with the joint-moments from the dry simulation and by reducing μ. Slip simulations with values of μ greater than the subject-specific peak required coefficient of friction (RCOF), an experimental measure of slip-resistant gait, resulted in only minor deviations in gait kinematics from the dry condition. In contrast, slip simulations run in environments characterized by $\mu <$ peak RCOF resulted in body kinematics that were substantially different from normal/dry gait patterns, more specifically greater knee extension and hip flexion angles were observed in the slip simulations. These findings imply the need for early and appropriate active corrective responses to prevent a fall in environments with μ values less than the peak RCOF.

1. Introduction

Falls account for more than 20% of fatal injuries in workers over the age of 65 years old (Agnew and Suruda, 1993; Leamon and Murphy, 1995; Courtney et al., 2001), and are often initiated by slips. The findings of the US National Health Interview Survey questionnaire in 1997 (Courtney et al., 2001) indicated that slipping was the most common triggering event, precipitating 43% of same level falls, followed by tripping (18%) and loss of balance (14%). Experimental gait studies conducted on contaminated slippery floors and using simulated slip protocols (e.g. base of support translations) have improved our understanding of the complex relationship between gait biomechanics and falls (see e.g. Cham and Redfern, 2001a; Pai and Iqbal, 1999; Tang et al., 1998). Two types of biomechanical variables (post-slip and initial conditions) may impact the magnitude of a balance loss induced by a slip and thus the resulting risk of fall. In moderate and severe slips that occur at or shortly after heel contact, an active response generated in the leading (slipping) leg is needed to prevent a fall (Cham and Redfern, 2001b). This response consists of a knee flexion moment and a hip extension moment and in general occurs about 150–200 ms after heel contact (Cham and Redfern, 2001b). While experimental studies have been useful to identify biomechanical variables that may impact slipping severity, it is unclear how these factors interact with each other and with environmental factors such as a reduced friction at the shoe–floor interface. This knowledge will provide the biomechanical reasons for one individual to slip and to recover while another slips and falls in the same environment; important information needed for the development of falls-related intervention programs (Redfern et al., 2001). This knowledge cannot be acquired through experiments alone. Whole body simulations of slipping are needed to meet these research needs.

Human gait models of varying complexity have provided insight into how the neuromuscular system functions during locomotion (Hurmuzlu, 1993a, 1993b; Neptune, 2000a; Neptune et al., 2000c, 2001; Pandy and Berme, 1989a, 1989b; Delp et al., 1990, 1996; Delp and Loan, 1995; Tashman et al., 1995; Tagawa and Yamashita, 2001; Ephanov and Hurmuzlu, 2002; Piazza et al., 1998, 2003; Piazza and Delp, 1996, 2001), permitted analysis of prosthetics and orthotics (Neptune et al., 2000c), and suggested techniques for

* Correspondence to: 3700 O'Hara St., 749 Benedum Hall, Pittsburgh, PA 15261, USA. Tel.: +1 412 648 7364; fax: +1 412 383 8788.
E-mail address: arm19@pitt.edu (A. Mahboobin).

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enhancing athletic performance (King and Yeadon, 2002; Yeadon and King, 2002). Although dynamic models have been used in a wide range of gait and neuro-rehabilitation research applications, there have been very few reports on the use of forward dynamic simulations in investigations of postural reactions generated in response to external perturbations during walking (Yang et al., 2008; Cordero et al., 2004; van den Bogert et al., 2002; Zhou et al., 2002). Modeling investigations described in these studies demonstrate how such an approach can play an important role in focusing therapeutic interventions on factors that contribute to an increased chance of recovering from external perturbations.

The purpose of this work was to create forward dynamic simulations of human walking both on a dry surface and with slipping. These simulations will be used to determine the effect of a systematic reduction in the coefficient of friction between the foot and the floor (μ) on the kinematics of walking during a brief period after heel contact before an active (voluntary) response is likely to play a major role. The potential findings of this work are significant for three reasons. First, the results of the simulations could be used to reduce the risk of slips and falls. Forward simulations that replicate slipping behavior will permit identification of the causal relationship between biomechanical factors and slip and fall risk by varying one simulation parameter and keeping other parameters unchanged, and such knowledge cannot be revealed through experiments alone. For example researchers cannot expose the same subject to multiple slips (with different μ values) because learning and anticipation effects would render subsequent trials useless (Cham and Redfern, 2002). Second, the findings may provide insights into the impact of active corrective moments needed to prevent a fall. Specifically, it is expected that as μ is reduced, body kinematics will diverge from normal (slip-resistant) gait. A greater difference in the body kinematics between the dry and slip simulations implies a greater need for an early and larger corrective response to prevent a fall. Third, the simulation results may have important implications in safety-related research, specifically in the design of slip-resistant shoe and flooring environments.

2. Methods

Normal walking kinematics and bilateral ground reaction forces were collected for two healthy subjects (Subject 1: male, age=25 yr, mass=69 kg, height=170 cm; Subject 2: female, age=22 yr, mass=55 kg, height=160 cm). Subjects walked at a self-selected speed along an 8.5 m long walkway. A Vicon (Vicon Peak—UK) motion capture system recorded three-dimensional motion data at a sampling rate of 120 Hz from 79 reflective markers placed on the body and shoes (Moyer et al., 2006). Ground reaction forces were recorded at a sampling rate of 1080 Hz and synchronized with the motion data. Following the normal walking, a glycerol solution was applied onto the force plate without the subject's knowledge to generate an unexpected slip at heel contact of the left foot (leading/slipping leg). Written informed consent approved by the Institutional Review Board of the University of Pittsburgh was obtained prior to any testing.

A two-dimensional musculoskeletal model, constrained to the sagittal plane, was developed in OpenSim (Delp et al., 2007) and used to obtain a kinematically-consistent set of joint angles and moments of normal gait using the measured motions and ground reaction forces. The model was composed of 10 segments (Fig. 1): head–arms–trunk (HAT), pelvis, left/right thigh, left/right shank, left/right rear-foot, and left/right toe. Revolute joints were used to represent the metatarsophalangeal joints, ankles, knees, hips, and a ''lumbar'' joint between the pelvis and HAT. The model had 12 degrees-of-freedom (DOF). The inertial properties for the body segments were drawn from Anderson and Pandy (1999, 2001). Subject-specific models were created by scaling the planar model in OpenSim to match each subject's anthropometry. Specifically, each body segment in the model was scaled based on relative distances between experimental and virtual marker locations, and mass properties of each segment was scaled proportionally such that the total mass of the subject is reproduced. Details of the scaling and inverse kinematic procedures were provided by Delp et al. (2007).

Moment-driven forward-dynamic simulations incorporating foot–floor interaction were created using SIMM/Dynamics Pipeline (MusculoGraphics, Inc.) and SD/Fast (PTC, Inc.). The topology of the OpenSim scaled model was defined in SIMM and Dynamics Pipeline was then used to create specialized C-language code

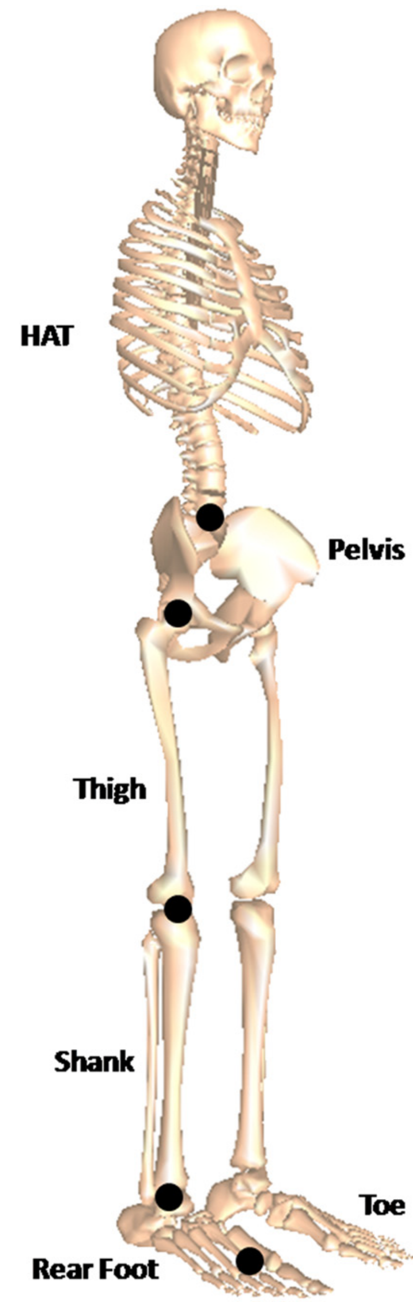

Fig. 1. Planar (2D) musculoskeletal model developed in OpenSim (Delp et al., 2007). The model is composed of 10 segments: head–arms–trunk (HAT), pelvis, left/right thigh, left/right shank, left/right rear-foot, and left/right toe. Revolute joints (●) were used to represent the metatarsophalangeal joints, ankles, knees, hips, and a ''lumbar'' joint between the pelvis and HAT. The model has 12 degrees-of-freedom.

that is compiled into an executable simulation along with model-specific C-libraries for kinematics and dynamics generated by SD/Fast. Contact between the feet and the floor was modeled using the spring-based contact that is based on the scheme described by Neptune et al. (2000b) and which includes friction. A quasi-Coulomb friction model was used in which highly viscous damping was applied when sliding velocities were small and friction forces proportional to normal forces ($F_f=\mu N$) were applied otherwise. Specifically, the vertical and horizontal force for each contact element i was calculated as

$$F_i^V = a_0(a_1 v_i^{a_2} + a_3 v_i^{a_4} \dot{v}_i^{a_5}) \quad (1)$$

$$\begin{aligned} F_i^H &= a_0 a_6 \dot{h}_i \qquad \text{if } |F_i^H| \le \mu F_i^V \\ F_i^H &= -\mu F_i^V sign(\dot{h}_i) \qquad \text{if } |F_i^H| > \mu F_i^V \end{aligned} \quad (2)$$

where F^V and F^H are the vertical and horizontal forces, respectively, a_0 is a scaling factor, $a_1, \ldots, a_5$ represent shoe-specific parameters, a_6 viscous damping coefficient for low sliding velocities, v the vertical deformation, $\dot{v}$and $\dot{h}$ the vertical and horizontal velocities, and μ the coefficient of friction set to 1.0. The values for a_0, a_2, a_3, a_4, a_5, and a_6 were 0.0684, 2.2, -16000, 0.8, 1.5, and 3000, respectively (Neptune et al., 2000b), and a_1 was scaled to each subject's body weight and was set to 1440000 (Subject 1) and 1450000 (Subject 2).

The hard-sole shoe used in our gait experiments was scanned (Fig. 2, top row) and incorporated in the model to aid in spring placement (Fig. 2, bottom row). The scanned shoe was first split into two parts, rear and front (Fig. 2, top row), translated to its most posterior and inferior point, then scaled to match the subject's actual shoe length. This step was followed by translating the shoe to the ankle and the toe coordinate systems, to have the rear and front part of the shoe correspond to the rear-foot and toe segments, respectively. Contact elements, spaced 2 cm apart (89 per foot for Subject 1 and 72 per foot for Subject 2, whose feet were smaller), were placed along the plantar surface of each foot segment. The shoe scanning provided us with a systematic way to include springs in our model and compare simulation results as a function of contact elements. Although the model was constrained to the sagittal plane, springs were placed in three dimensions so that the model could be used in more complex 3D simulations of walking/slipping in the future. In the simulations presented here, we are only concerned with the vertical and anterior–posterior ground reaction forces.

A parameter optimization was performed in which the set of joint moments was found that minimized J, the sum of the squares of the differences between simulated (sim) and measured (exp) joint rotations (JR) and ground reaction forces (GRF),

$$J = \sum_i w_i^{JR}(\mathrm{sim}_i^{JR} - \exp_i^{JR})^2 + \sum_j w_j^{GRF}(\mathrm{sim}_j^{GRF} - \exp_j^{GRF})^2 \quad (3)$$

where w^{JR} and w^{GRF} represent the corresponding weights for the joint rotations (i=1,2, …, 8; representing pelvis tilt, left/right hip, knee, and ankle, and lumbar joints) and ground reaction forces (j=1, …, 4; representing left/right shear and normal forces), respectively. These weights were adjusted by trial and error, between parameter optimization runs, depending on which simulated joint rotation or ground reaction force component differed most from its experimental value. Joint rotation and ground reaction force weights ranged from 80,000–120,000 and 0.08–5.12, respectively, and differed between each optimization run. Prior to running the optimization, the measured (exp) joint angles (sampled at 120 Hz) and ground reaction forces (sampled at 1080 Hz) were truncated from heel contact of the leading leg to 200 ms after heel contact. Cubic splines were then fit to the data for smoothing and resampling (100 Hz) purposes, resulting in 20 discretized points evenly spaced in time. The joint moments used in the optimization were the bilateral hip, knee, and ankle moments, and the lumbar extension moment. Each of the seven joint moments was discretized into 20 values evenly spaced in time (Pandy et al. 1995) between t=0 and 0.19 s, giving 140 parameters. The optimization was accomplished using a hybrid of particle swarm (Kennedy and Eberhart, 1995) and Nelder–Mead downhill simplex (Press et al., 1988) methods. In this implementation, downhill simplex optimizations were performed after every 50 particle swarm iterations using the global best particle and five randomly selected particles (from 30 total particles). Maximum instantaneous errors of less than 10°, 25 N, and 100 N were judged to represent acceptable qualitative agreement between the simulated and measured joint rotations, horizontal ground reaction forces, and vertical ground reaction forces, respectively. When these errors were not exceeded following a single parameter optimization run (5000 particle swarm iterations), no further parameter optimization runs were performed.

Normal gait was simulated from the leading/slipping leg heel contact (0 ms) to 190 ms after heel contact. This duration was selected because active corrective responses during slipping are triggered about 150–200 ms after heel contact (Cham and Redfern, 2001b). Before initiating the optimization to determine the joint moment histories, a preliminary optimization was performed to obtain a starting configuration that matched the initial ground reaction forces.

Finally, after establishing a simulation that satisfactorily tracked normal walking, slip was simulated by reducing the frictional forces applied to the leading foot (i.e., by reducing μ). The value of μ was decreased systematically (in decrements of 0.01 from the initial value of μ=1.0) until deviation from the dry pattern emerged. The applied joint-moments remained the same as in the normal (dry) gait simulation. To assess the slip simulations, we quantified the ratio of the shear to normal forces (Redfern et al., 2001). Under dry conditions, this quantity is referred to as the required coefficient of friction (RCOF) and it has been suggested that for friction values below the peak RCOF, there is a greater potential for slipping (Redfern et al., 2001).

3. Results

The optimization of normal/dry walking resulted in joint moments that reproduced salient features of the experimentally collected data (Fig. 3). Overall, the root-mean-square errors (RMSE) between the simulated and experimental joint rotations for Subjects 1 and 2 were less than 6° and 3°, respectively. The RMSE between the simulated and measured shear and normal ground reaction forces, for both the leading/slipping and trailing leg, were less than 18 and 33 N, for Subject 1, and 25 and 21 N, for Subject 2, respectively. Of particular interest are the leading/slipping leg hip and knee joints (Fig. 3, top two panels), since it is believed that the leading/slipping leg hip and knee joints are important in arresting a sliding motion of the leading foot and in arresting the vertical descent of the body during a balance loss (Cham and Redfern, 2001b). The RMSE between the simulated and

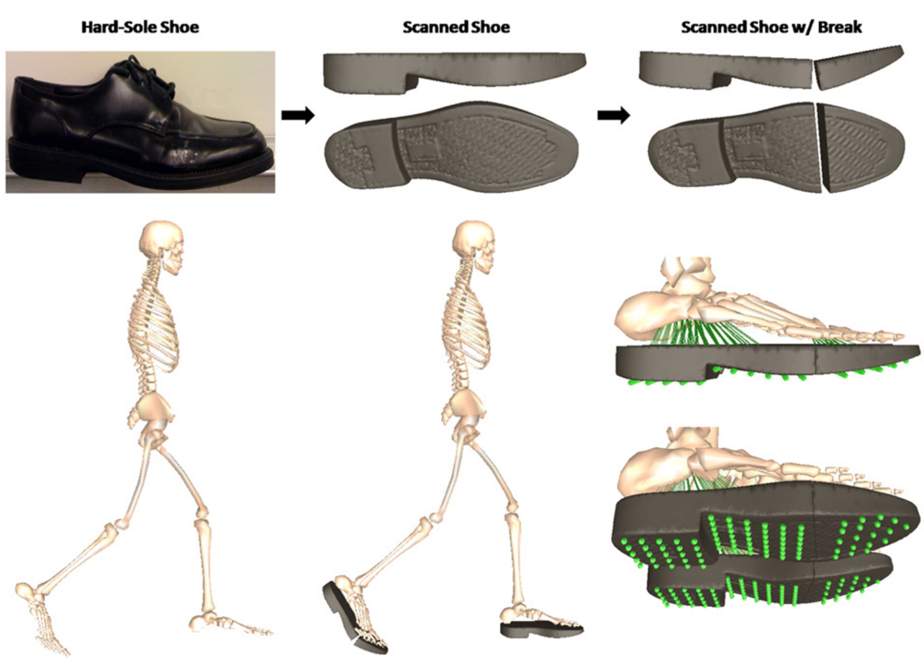

Fig. 2. Foot–floor interaction model. The top panel shows the hard-sole shoe and its scanned version. The shoe was split into two parts, rear and front, to correspond with the rear-foot and toe segments, respectively. Contact elements (89 per foot for Subject 1 and 72 per foot for Subject 2 [smaller feet]), spaced 2 cm apart, were placed along the plantar surface of each foot segment, following the contour of the shoe sole.

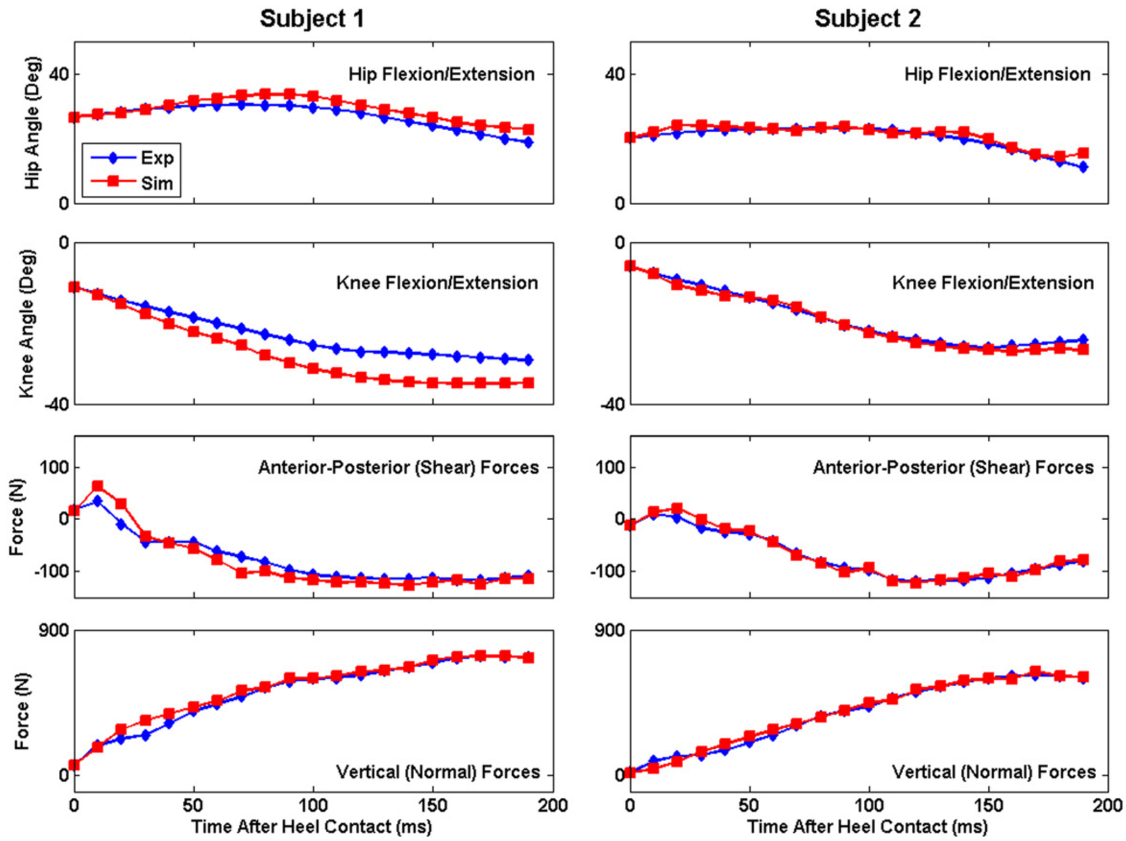

Fig. 3. Normal/dry joint angle and ground reaction force comparisons between experimental (diamond) and simulation (square) data, where the top two panels show the leading/slipping leg hip and knee flexion/extension joint angles and the bottom two panels show the anterior–posterior (shear) and vertical (normal) ground reaction forces for Subject 1 (left column) and Subject 2 (right column), respectively.

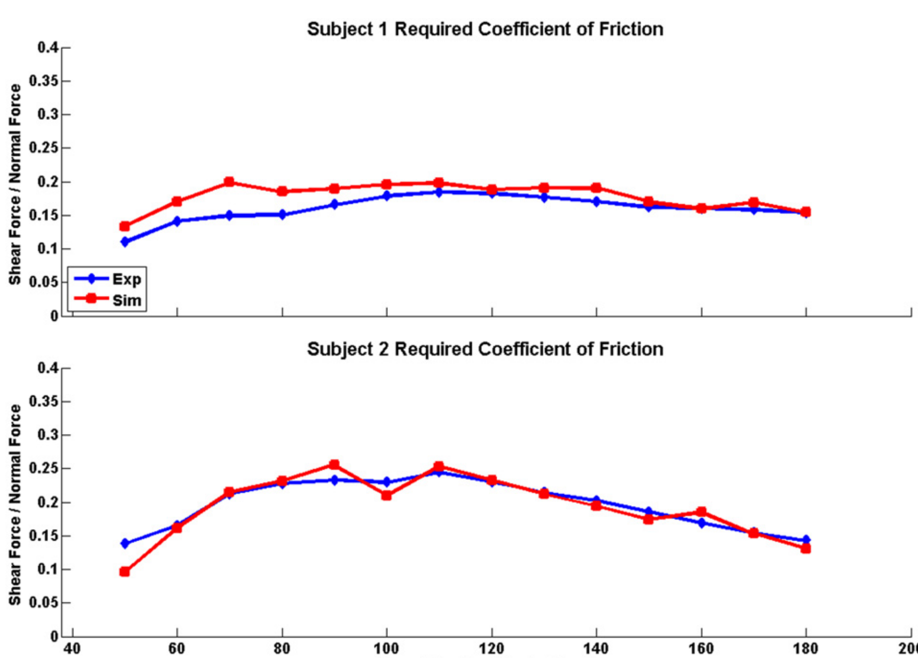

Fig. 4. Computed experimental and simulated required coefficient of friction (RCOF), calculated using the normal/dry ground reaction forces, for Subject 1 (top panel) and Subject 2 (bottom panel).

experimental hip and knee joint angles of the leading/slipping leg, for Subject 1, were less than 2.4° and 5.1°, respectively. The leading/slipping leg ground reaction force RMSEs were less than 15 N for the shear force and 30 N for the normal force. For Subject 2, the RMSE between the simulated and experimental hip and knee joint angles of the leading/slipping leg were less than 1.4°

and 1.0°, respectively, and the leading/slipping leg ground reaction force RMSEs were less than 7 N for the shear force and 21 N for the normal force.

The experimental and simulated peak RCOF patterns were very similar (Fig. 4). The experimental and simulated peak RCOF computed for Subject 1 (Fig. 4, top panel) was at 0.18 and 0.19, respectively. Subject 2's (Fig. 4, bottom panel) peak RCOF was at 0.24 (experimental) and 0.25 (simulation).

As expected, during the early stance period (~190 ms) following heel contact, the difference in walking kinematics between the normal gait and slip simulations increased as μ was decreased (Fig. 5). Specifically, applying the required joint moments needed in normal/dry walking in the slip simulations with values of $\mu \geq 0.18$ (for Subject 1) and 0.21 (for Subject 2), resulted in only a minor deviation in gait kinematics from the normal/dry condition (Table 1), based on the overall RMSE (computed from 0 to 190 ms) of 0.92° and 2.8° (for Subject 1) and 0.5° and 3.1° (for Subject 2) for the hip and knee angles of the leading/slipping leg, respectively. In contrast, the slip simulations indicated that walking with normal/dry joint moments in environments characterized by μ values less than 0.18 and 0.21 results in kinematics that will be substantially different from normal/dry gait patterns (Table 1). Additionally, as μ was reduced, the slip simulations resulted in more rapid deviation from normal gait kinematics in stance (Fig. 5).

There was general agreement between the experimental and simulated slip joint angle results (Fig. 6), especially up to about 150 ms after heel contact, where both the experimental and simulation trajectories show similar patterns for low values of μ. The divergence observed between the simulated and experimental slips after 150 ms is due to activation of corrective responses in the leading/slipping leg in the latter case to prevent a fall. There was also general agreement between the slip experimental and simulated normal (dry) joint moments (Fig. 7). In the slip experiment, the application of a glycerol contaminant prevented the generation of normal level shear forces prior to 150 ms, which may partially explain the difference in the joint moments between the experimental findings (a real slip) and the simulated moments generated using data obtained in a dry environment (Fig. 7).

4. Discussion

We introduced a method in which a simulation of normal walking was created to determine the joint kinematics, ground reaction forces, and applied joint moments for an interval of

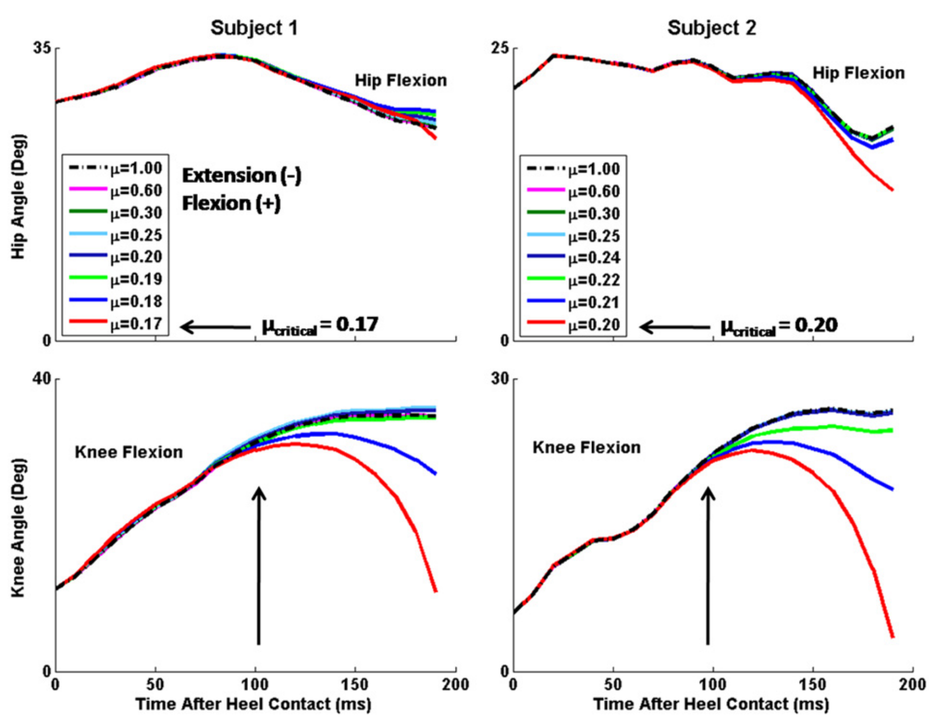

Fig. 5. Leading/slipping hip and knee flexion angle comparisons between normal/dry walking (black dotted-dashed line) and slip simulations for Subject 1 (left column) and Subject 2 (right column). Note that for values of $\mu \geq 0.18$ (Subject 1) and 0.21 (Subject 2), the model is capable of producing consistent gait patterns that are similar to the dry condition, only utilizing normal gait moments. At the later stages of stance, the kinematics on slippery floors diverge from the dry condition (μ less than 0.18 and 0.21), indicating the need for active corrective moments. The vertical arrows indicate the onset of divergence in the knee joint, as compared to the hip joint diverging later in the stance phase as μ is reduced. This phenomenon has also been observed experimentally (Cham and Redfern, 2001b).

Table 1
Overall root mean square error (RMSE) and maximum absolute difference of the hip and knee joint angles of the leading/slipping leg for Subjects 1 and 2.

	Subject 1			Subject 2		
	$\mu \geq 0.18$	$\mu < 0.18$		$\mu \geq 0.21$	$\mu < 0.21$	
	RMSE (deg.)	RMSE (deg.)	Max abs diff (deg.)	RMSE (deg.)	RMSE (deg.)	Max abs diff (deg.)
Hip	0.92	0.5	1.5	0.5	2.1	7.4
Knee	2.8	7.4	24	3.1	7.3	23

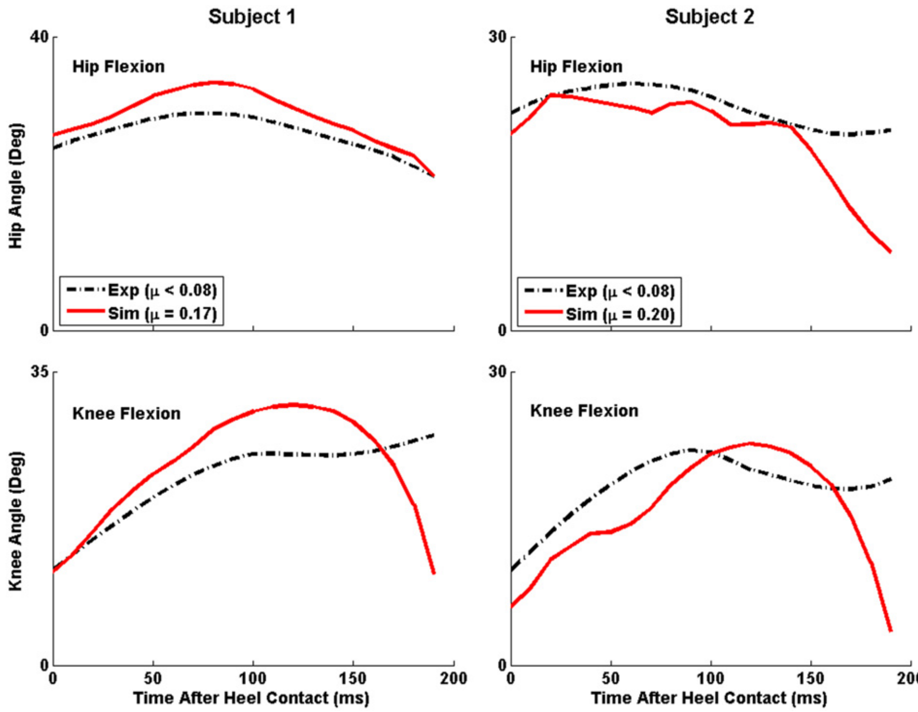

Fig. 6. Leading/slipping hip and knee flexion angle comparisons between experimental slipping (black dotted-dashed line) and slip simulations for Subject 1 with μ=0.17 (left column) and Subject 2 with μ=0.20 (right column). Note that at about 150 ms after heel contact, the experimental hip and knee joint angles diverge from their simulated counterparts, implying a need for applying active corrective moments to prevent a fall.

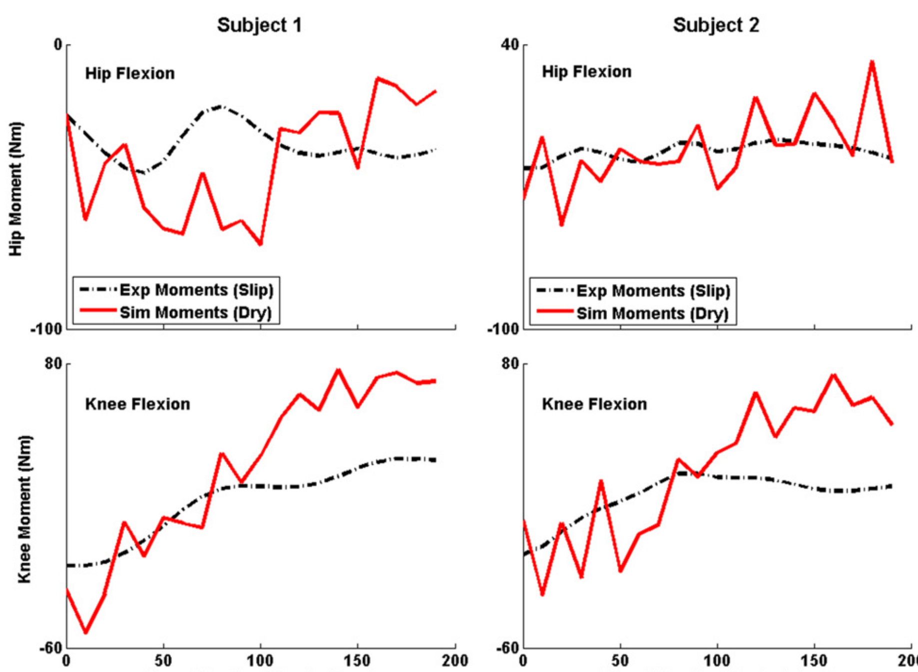

Fig. 7. Leading/slipping hip and knee flexion moment comparisons between experimental slipping (black dotted-dashed line) and normal (dry) simulations for Subject 1 (left column) and Subject 2 (right column).

190 ms after heel contact of the leading/slipping leg. Simulations of the slips of two subjects were performed using the joint moments determined for the dry walking simulation but with reduced foot–floor coefficients of friction.

Simultaneous tracking of joint angles and measured ground reaction forces can be a difficult problem, but reasonable results were achieved in the present study using a simplified planar model with spring-based foot–floor contact in our normal walking simulations. Most previous efforts at modeling foot–floor contact (e.g., Neptune et al., 2000b; Anderson and Pandy, 1999; Yang et al., 2008) have utilized much simpler spring node arrays on the plantar surface of the foot. More complex models are probably necessary to accurately represent the ground reaction forces at heel contact to the degree necessary for studying slipping behavior. Incorporating the scanned shoe in the present study likely contributed to the good agreement between the measured and simulated ground reaction forces that we have found (Fig. 3, bottom two panels). The scanned shoe allowed us to place springs at locations that closely approximate the contour of the bottom surface of the shoe. We found that placing the springs in this fashion, as opposed to the simpler spring placements mimicking a flat shoe sole, and having a sufficient number of springs, was an effective way to obtain good tracking results as shown here. Preliminary simulations for the two representative subjects with fewer spring elements (15 per foot for Subject 1, and 16 per foot for Subject 2), spaced 4 cm apart (as opposed to 2 cm), were conducted (Fig. 8). As evident from the plots, the tracking was not as effective (joint rotation RMSEs less than 12° and 6°, for Subject 1 and 2, respectively. And ground reaction force RMSEs less than 42 N (shear) and 124 N (normal), for Subject 1, and 36 N (shear) and 60 N (normal), for Subject 2) when compared to results presented in Fig. 3.

The slip simulation findings imply the need for early and appropriate active corrective responses to prevent a fall, especially in slippery environments with μ values less than the peak RCOF. This need was evident when comparing the experimental and simulated slips (Fig. 6), where approximately ~150 ms after heel contact of the leading/slipping leg the observed divergence between the experimental and simulated trajectories indicates need for active corrective responses to prevent a fall. The onset of active corrective moments are generally believed to be initiated ~150–200 ms after heel contact (Cham and Redfern, 2001b). This is also confirmed with EMG data (Chambers and Cham, 2007).

An interesting finding of this study was the agreement between the simulation peak RCOF and the critical μ value obtained for each subject. Experimentally, it has been shown that the peak RCOF for non-slip gait ranges from 0.17 to 0.22 (Redfern et al., 2001). Our simulation findings agree with these results, further validating our simulation approach. Specifically, the correspondence between the peak RCOF and the critical μ identified in our slip simulations supports the validity of our simulation results.

A limitation to our approach is the use of a planar model rather than a three-dimensional model. More complex models will be needed to fully understand slipping behavior, as experimentally measured postural recovery responses from slips are not restricted to the sagittal plane. Another limitation derives from the assumption that the upper and lower arms are fixed to the trunk, because of the role arms play in fall recovery. An additional limitation to be considered is modeling the head as part of the trunk to capture contributions of head movements that are distinctly different from the trunk during walking. Future work on these modeling methods will include adding complexity to address these limitations, substituting the manual weight adjustment used in our optimizations with an automated scheme (Reinbolt et al., 2008), as well as including muscles and the use of deterministic, gradient-based optimization methods to improve tracking once the currently used stochastic methods have produced a solution near the global minimum. Use of such methods could result in smoother trajectories not obtained here, specifically in the joint moment histories (Fig. 7).

The results of the simulations have the potential to reduce the risk of slips and falls in the workplace. The findings can be used to

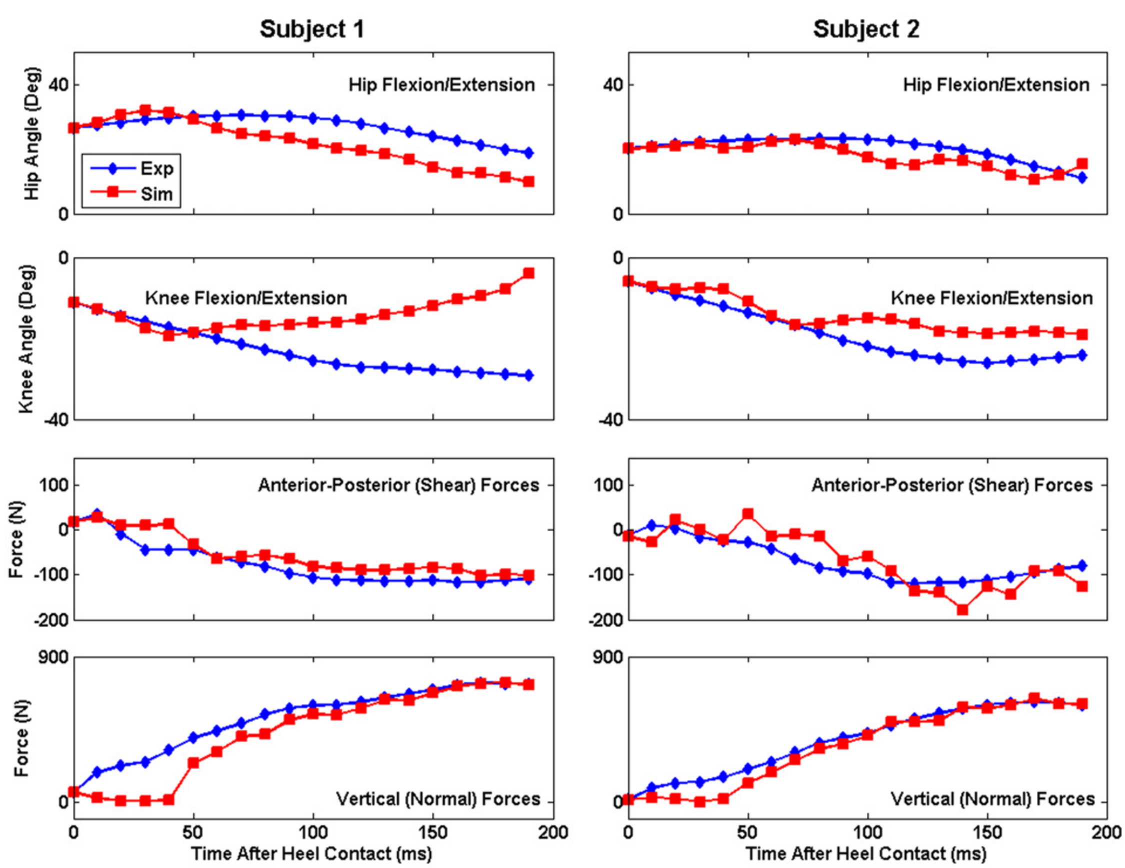

Fig. 8. Normal/dry joint angle and ground reaction force comparisons between experimental (diamond) and simulation (square) data, for Subject 1 (left column) and Subject 2 (right column) using fewer spring elements (15 per foot for Subject 1, and 16 per foot for Subject 2). The top two panels show the leading/slipping leg hip and knee flexion/extension joint angles and the bottom two panels show the anterior–posterior (shear) and vertical (normal) ground reaction forces for Subject 1 (left column) and Subject 2 (right column), respectively.

design occupational falls-related prevention programs focused on improving human and biomechanical factors that have been shown through simulations to increase the risk of falls. For example, to decrease the risk of slips/falls, improving the ability to generate knee and hip joint moments rapidly may be more important than maximizing peak strength. The gait simulations of slipping can also play a role in the design of the workplace environment (e.g., in the selection of appropriate shoe–floor material characteristics that meet biomechanical requirements of walking without slipping). There is a trade-off between the frictional characteristics of a floor and the forces and stresses that the floor would impose on the joints. Maximizing floor friction may promote joint injuries, and the amount of friction needed to minimize the risk of slips/falls is unclear. This information would depend on the biomechanical requirements of a successful slip-recovery reaction, which could be investigated using the proposed modeling techniques.

In conclusion, our simulation results obtained here encourage the use of multi-DOF whole body models with foot–floor interaction to study perturbed walking with the hope of reducing falls in older adults and other clinical populations. Such approach would complement experimental gait studies and help disentangle confounding factors often difficult to understand in experimental work alone. We believe that the agreement obtained between the experimental and simulated data reported in this work is, to a large extent, due to use of subject-specific modeling, incorporating the scanned shoe in the contact model, and initial conditions.

Conflict of interest

There are no conflicts of interest.

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