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Role of individual lower limb joints in reactive stability control following a novel slip in gait

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ABSTRACT

Instability after slip onset is a key precursor leading to subsequent falls during gait. The purpose of this study was to determine the impact of reactive muscular response from individual lower limb joints on regaining stability control and impeding a novel and unannounced slip during the ensuing single-stance phase. Ten young adults' resultant moments at three lower limb joints of both limbs, initially derived by an inverse-dynamics approach from empirical data, were optimized to accurately reproduce the original motion before being applied as input to the control variables of their individualized forward-dynamics model. Systematic alteration of the moments of each joint caused corresponding changes in the displacement and velocity of the center of mass (COM) and base of support (BOS) (i.e. their state variables, x_{COM}, $\dot{x}_{COM}$, x_{BOS}, $\dot{x}_{BOS}$), and in the COM stability. The model simulation revealed that these joints had little influence on $\dot{x}_{COM}$ but had substantial impact on $\dot{x}_{BOS}$ reduction, leading to improve the COM stability, mostly from knee flexors, followed by hip extensors, of the slipping limb. Per unit reactive *increase* in normalized knee flexor or hip extensor moments and per unit reactive *reduction* in commonly observed plantar–flexor moments could lead to as much as 57.72 ± 10.46 or 22.33 ± 5.55 and 13.09 ± 2.27 units of *reduction* in normalized $\dot{x}_{BOS}$, respectively. In contrast, such influence was negligible from the swing limb during this period, irrespective of individual variability.

1. Introduction

Falls are a major cause of injury and even death in adults 65 years or older. In the United States, over 1.85 million older adults were treated in the emergency room for fall-related injuries in 2004 (Bieryla et al., 2007). Slip-related falls account for about 25% of all falls among older adults (Holbrook, 1984) and frequently cause hip fracture that can have devastating consequences (Kannus et al., 1999). A better understanding of the mechanisms underlying the control of stability during slip-related falls will undoubtedly be an important step towards the prevention of such injuries and reduction of the cost resulted from the slip-related falls.

One of the fall prevention approaches may rely on the adaptive improvements of an individual's control in dynamic stability following the onset of perturbation (Pai, 2003; Pai and Bhatt, 2007). The center of mass (COM) stability, which can be measured by the shortest distance from the relative motion states (i.e. the instantaneous displacement and its velocity) between the COM and its base of support (BOS) to the dynamic stability limits (Yang et al., 2008b), plays an important role in recovery from a forward slip (Bhatt et al., 2006; Pai, 2003; Pavol and Pai, 2007; You et al., 2001). Four state variables, i.e. the displacement of COM and BOS (x_{COM} and x_{BOS}) and their corresponding velocity ($\dot{x}_{COM}$ and $\dot{x}_{BOS}$), therefore directly dictate the stability during a slip. Empirical evidence indicates that the velocity of the slipping foot (i.e.$\dot{x}_{BOS}$) is a key factor affecting the recovery outcome following a slip (Bhatt et al., 2006; Cham and Redfern, 2002; Lockhart et al., 2003; Strandberg and Lanshammar, 1981). Yet, the relationship of these four state variables and the COM stability has not been systemically analyzed.

It has been demonstrated that adaptive control of stability can improve the slip recovery outcome to such an extent that successful feedforward control can alleviate or even completely eliminate the need for reactive correction after the onset of a slip (Bhatt et al., 2006; Pai et al., 2000, 1998, 2003). Conversely, following a novel and unannounced slip induced in gait, the COM stability deteriorates rapidly and severely. Such deterioration continues after slip onset during the first ~180 ms of double-stance and during the subsequent ~100 ms single-stance phase, in which only the slipping foot provides the BOS. Little is known how the COM stability is controlled during this crucial period.

The resultant joint moments, especially those from the lower limbs, are responsible for the control of locomotion (Winter, 1980) and they result directly from muscle activation that is governed by a descending motor program initiated from and modulated by

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various motor centers of the central nervous system. Through comparing the results of slipping trials and those of regular walking trials derived from an inverse-dynamics approach, it has been postulated that increased knee flexor and hip extensor moments at stance limb might be two primary reactive responses required to stabilize human body and to avoid a slip-related fall in gait (Cham and Redfern, 2001). Such comparisons do not in itself reveal the direct causal effect.

A combination of inverse-dynamics analysis and analytical manipulation of a forward-dynamics model to simulate *an individual's* performance may reveal the mechanistic underpinning the COM stability control. By systematically altering the joint moments, one at a time while keeping initial motion state of body segments and the other joint moments constant during a forward-dynamics simulation, the exclusive causal relationship between each individual joint moment and stability control may be quantitatively evaluated. Nonetheless, this will lead to a classic paradox here. Namely, the joint moments derived from an inverse-dynamics approach often cannot reproduce original motion when applied as an input to a forward-dynamics model, presumably resulting from error inherent to kinematic and ground reaction force (GRF) data collection (Kuo, 1998). Recent attempts have been taken to reduce this kind of inherent error in the joint moments (Kuo, 1998; Neptune et al., 2001). Such approach is yet to be applied to explore causal relationship between the joint moments and the reactive control of the COM stability.

The purpose of this study was to determine the impact of reactive muscular response from individual lower limb joints on regaining stability and impeding a novel and unannounced slip during the ensuing single-stance phase. This objective was achieved by systematically altering the optimally matched lower joint moments in a forward-dynamics simulation based on personalized individual human models and their actual recorded performance during single-stance phase.

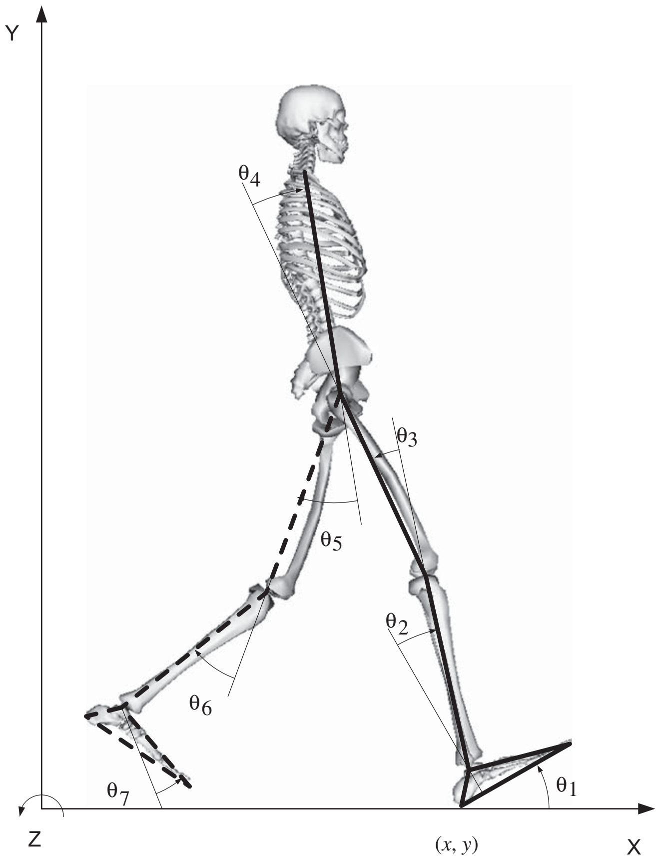

Fig. 1. Schematic of the 7-link, 9-degree-of-freedom, sagittal-plane model of the human body. The vector $\mathbf{q}=[x,y,\theta_1,\theta_2,\ldots,\theta_7]$ represents the generalized coordinates of the model. Coordinates x, y, and θ_1 specify the position and orientation of the stance (right) foot, which is the base segment of the model following the swing (left) foot liftoff after the slip onset, with reference to the inertial reference frame (X, Y, Z). During the single-stance phase, the area under the right foot is the base of support (BOS) of the human model. Joint angles θ_i (i=2, 3, …, 7) correspondingly specify the angles of the ankle, knee, hip of the stance limb (sold line) and the hip, knee, and ankle of the swing limb (dashed line). The segment lengths of an individual model are calculated from the relative distance between pairs of joint centers measured for that individual subject. The location of the center of mass for each segment shown as the half-shaded circle, as well as its moment of inertia are estimated based on the subject's body mass and the measured segment length (de Leva, 1996). The positive X-axis is in the direction of forward progression and the positive Y-axis is upward. Positive joint rotation is along the positive Z-axis (counterclockwise) for the stance limb (solid line), and its sign is reversed (clockwise) for the swing limb (dashed line).

2. Methods

The data of ten young adults were randomly selected from an existing database collected during their first encounter of a novel and unannounced slip while walking (Bhatt and Pai, 2009; Bhatt et al., 2006). The mean $\pm$ SD body height and mass were 169.4 $\pm$ 7.0 cm and 64.7 $\pm$ 15.5 kg, respectively. All subjects have given written informed consent to the experimental protocol approved by the Institutional Review Board. Every one took at least 10 unperturbed walking trials at their self-selected speed in which a passively movable platform was locked and mounted on a low-friction linear bearing on a supporting frame (Yang and Pai, 2007), while they were only told that a slip would be possible (Bhatt et al., 2006). No information was given as to where, when, and how a slip would occur when this slip was actually induced with the release of this platform that was camouflaged by similar decoy structures. In response to this novel and unannounced slip, all subjects experienced backward balance loss by taking a recovery step that landed posterior to the slipping foot. Full-body kinematic and kinetic (i.e. GRF) data were collected for this trial (Bhatt et al., 2006) and were included in the following two stages of the present study.

The objective of the first stage was to develop individual human models (Fig. 1) and to derive the resultant joint moments, τ, of the lower limb, first with inverse-dynamics formulation, and then with simulated annealing optimization routine. The resultant joint moments were normalized by the product of the body mass, bm; the gravitational acceleration, g; and the body height, bh. These *optimally matched moments* could best replicate, or best fit, each subject's measured body kinematics and kinetics during gait-slip experiments (Yang and Pai, under review). Due to the perturbation induced upon the slip trials which alters the kinematics and kinetics of the body segments, the optimally matched moments differ from the joint moments during regular walking trials (Schwartz et al., 2008). Such differences have been also found by other studies (Cham and Redfern, 2001; Ferber et al., 2002). The objective of the second stage was to apply this *individualized* model to explore the relationship of these joint moments with these four state variables and with the COM stability. Specifically, with input from experimentally derived initial segment motion state at left liftoff, the time profile of the optimally matched moment during single-stance phase served as the control variables of forward-dynamics model for this individual subject. The control variables were altered systematically, one joint at a time, by adding or subtracting a fixed increment of 10^{-4} ($bm \times g \times bh$) from their optimally matched moments throughout the single-stance phase (Fig. 2). This process of augmentation would be terminated until the point at which the left foot contacted the ground before the termination of the perturbed simulation, or at which any joint angle from simulation became anatomically unrealistic, i.e. when it begins to exceed one stand deviation beyond average range of motion for this particular joint (Yang and Pai, under review).

For each alteration in τ, we determined its effect by computing the changes in four state variables (i.e. x_{COM}, $\dot{x}_{COM}$, x_{BOS}, $\dot{x}_{BOS}$) and in the COM stability (s) at the end of the single-stance phase. Both x_{COM} and x_{BOS} were normalized to foot length, l_{BOS}, while $\dot{x}_{COM}$ and $\dot{x}_{BOS}$ were normalized to $\sqrt{g \times bh}$. Using these ten subjects, the ratios of the leg length and l_{BOS} to bh are 51.58 $\pm$ 1.19% (R^2=0.82, $p<0.001$) and 17.43 $\pm$ 0.33% (R^2=0.81, $p<0.001$), respectively. As afore-mentioned, the instantaneous measurement of s was calculated as the shortest distance from the relative COM motion state (i.e. $x_{COM/BOS}$ and $\dot{x}_{COM/BOS}$) to the threshold against backward balance loss under slip condition in gait (Yang et al., 2008a). The model simulation in this previous study predicts that based on anatomical and physiological limitations and environmental constraints, a backward balance loss must occur when the COM state is located below the threshold ($s<0$). Greater stability above the threshold ($s \geq 0$) means that a person will less likely experience backward balance loss because the forward COM momentum is sufficient to prevent that from happening (Pai et al., 2003).