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ROLE OF STABILITY AND LIMB SUPPORT IN RECOVERY AGAINST A FALL FOLLOWING A NOVEL SLIP INDUCED IN DIFFERENT DAILY ACTIVITIES

Feng Yang, **Tanvi Bhatt**, and **Yi-Chung Pai**
Department of Physical Therapy University of Illinois at Chicago Chicago, IL 60612

Abstract

The purpose of this study was to determine whether stability and limb support play a similar role in governing slip outcome in gait-slip as in sit-to-stand-slip, and whether such prediction could also be derived based on measures of these variables during regular, unperturbed movements. Fifty-three and forty-one young subjects all took one recovery step following an unannounced, novel, forward slip induced in gait-slip and in sit-to-stand, respectively. Logistic regression was used to predict recovery outcome based on *preslip* and *reactive* measures of stability and limb support across tasks. Following slip onset, all subjects in both tasks experienced rapid decay in stability and limb support (indicated by a hip descent), leading to some actual falls that could not have been predicted from regular, preslip walking. Immediately before recovery step touchdown, stability and limb support could together best predict 88.9% and 100% falls respectively for gait-slip and sit-to-stand-slip. Because of differences in the execution of the recovery step, stability became a better predictor of fallers in sit-to-stand-slip than in gait-slip after recovery limb touchdown. Recovery steps were highly effective in restoring stability, regardless of outcome and task. The predictive strength of stability diminished in gait-slip or reduced in sit-to-stand-slip after touchdown, while limb support remained able to differentiate fallers from those who recovered in both tasks. When slip-induced instability was combined with inadequate limb support, falls were nearly inevitable in both tasks.

Keywords

slip; gait; sit-to-stand; assessment; volitional; triggered response; stepping; generalization; fall prediction

INTRODUCTION

Slip-related falls accounts for about 25% of all falls, which often lead to grave consequences such as hip fracture among older adults (Mathers and Weiss, 1998; Stevens et al., 2006). It is therefore imperative to understand the causes contributing to falls (Bentley and Haslam,

Corresponding author: (Clive) Yi-Chung Pai, PhD Department of Physical Therapy University of Illinois at Chicago 1919 West Taylor St., room 426 (M/C 898) Chicago, Illinois 60612, USA Tel: +1-312-9961507 Fax: +1-312-9964583 cpai@uic.edu.

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1998; Carpenter et al., 1999), to develop effective risk assessment tools and interventions for reducing their incidence. Many falls occur during transitions such as sit-to-stand (Rubenstein et al., 1994), and this task has often been a key component of fall risk assessment (Berg et al., 1992; Tinetti, 1986). It is unclear, however, whether findings derived from the performance of sit-to-stand are similarly applicable to walking. Particularly, it is unknown whether there are general differences in fall-resisting mechanisms during sit-to-stand-slip and in gait-slip, or whether differences between the two tasks during unperturbed, volitional performance could be equally revealing (Berg et al., 1992; Tinetti, 1986).

The center of mass (COM) motion state (*i.e.*, its position and velocity) assessed during *unperturbed* (preslip) gait may have limited accuracy predicting balance loss resulting from a perturbation (Bhatt et al., 2006b). However, the COM motion state with respect to the base of support (BOS) assessed during perturbation could predict recovery from slips (Bhatt et al., 2006a; Bhatt et al., 2006b; Pai et al., 2003). Avoiding balance loss or a fall after a slip is dependent upon *reactively* retarding the forward motion of the slipping foot (Brady et al., 2000; Cham and Redfern, 2001; Gronqvist et al., 2001; Lockhart et al., 2003; Redfern et al., 2001; Strandberg and Lanshammar, 1981) and rapidly lowering the recovery limb to the ground posterior to the slipping foot (Marigold et al., 2003; Tang and Woollacott, 1998). For instance, during sit-to-stand-slip, arresting falls depend on *both* dynamic stability and adequate limb support, characterized by feasible stability boundary and hip motion (Pai et al., 2006; Pavol and Pai, 2007). Instability and inadequate limb support account for 97% of falls during the event of sit-to-stand-slip (Pavol and Pai, 2007).

It is reasonable to speculate that recoveries from forward slip during sit-to-stand trials might be different from those resulting from slip-perturbed gait. For instance, gait-slip is initiated from an asymmetrical bipedal support position, while sit-to-stand-slip from a symmetrical bipedal position. Such differences in initial body segment motion state could result in different recovery stepping for the two tasks. In gait-slip, a recovery step typically travels forward following slip onset, but moves backward in sit-to-stand-slip. Nonetheless, there are also noticeable similarities between both tasks in response to a forward slip. The feasible stability region against slip-induced backward balance loss as predicted based on a 2-link model representing bipedal-symmetric sit-to-stand movement (Pai and Iqbal, 1999) is very similar to that based on a 7-link walking model (Yang et al., 2008a). Further, repeated-slip exposure reduces backward balance loss risk by similarly increasing feedforward control of the COM state stability in both tasks (Bhatt et al., 2006b; Pai et al., 2003). It remains possible that the control of stability and limb support could both play a dominant role in differentiating fallers from those who recovered for the first unexpected slips during gait as they do in sit-to-stand (Pavol and Pai, 2007).

The purpose of this study was to determine whether stability and limb support play a similar role in governing recovery outcome in gait-slip as in sit-to-stand-slip, and whether such prediction could also be derived based on measures of these variables during regular, unperturbed movements. We postulate that regardless of the differences in task objectives, during sit-to-stand and in gait, the reactive control of both stability and limb support assessed after an unannounced, novel slip would provide good prediction of a subsequent fall; this would be more accurate than such prediction made before slip onset during the unperturbed part of movement, for both tasks.

METHODS

This section describes two different sets of experiment. The first experiment included sixty-seven young subjects (35 females, age: 26 ± 5years; mass: 63 ± 13kg; height: 1.69 ± 0.09m) who experienced unannounced slip-perturbation induced in gait, while the second included another sixty young individuals (44 females; age: 25 ± 5years; mass: 67 ± 14kg; height: 1.69 ± 0.10m) exposed to unannounced slip-perturbation during sit-to-stand movement. There was no difference in demographics between the two experiments. Each individual had given informed consent before participating in the experiment.

Similar experimental protocols and setup were adopted in both tasks (Table 1). Specifically, an unannounced slip-perturbation was induced while subjects walked or stood from a stool respectively in gait-slip and sit-to-stand-slip experiments. All subjects were only told that a slip may occur sometime during their repeated task performance. The slip-perturbation was generated by releasing moveable platforms (Table 1) (Pavol and Pai, 2007;Yang and Pai, 2007). A full-body harness system was employed for subject protection. A load-cell measured the force exerted on the harness.

The motion capture systems (Motion Analysis, CA, for gait-slip, and Peak Performance, CO, for sit-to-stand-slip) were employed to collect data from 26 retro-reflective markers placed on the body to create a 13-segment model for COM kinematic calculation (de Leva, 1996). Marker paths were low-pass filtered at marker specific cut-off frequencies ranging from 4.5 to 9 Hz using fourth-order Butterworth filters. The load-cell signal and the ground reaction force were recorded at 600 Hz and synchronized with the motion system.

Stability and hip motion were analyzed for both tasks. Stability was the shortest distance from the instantaneous COM state to thresholds against backward balance loss (Pai and Iqbal, 1999; Yang et al., 2008a; Yang et al., 2008b). Outside the threshold, balance recovery is theoretically impossible without reestablishing a new BOS. The COM position, $X_{COM/BOS}$, and velocity, $\dot{X}_{COM/BOS}$, were calculated relative to the posterior border of BOS. The hip height, $Z_{hip}(t)$, and its velocity $\dot{Z}_{hip}(t)$ were defined as,

$$\begin{aligned} Z_{hip} &= \left(h_{r,hip}+h_{l,hip}\right)/2 \\ \dot{Z}_{hip} &= \frac{dZ_{hip}(t)}{dt} \end{aligned}$$

where, h is the height of the hip center. Subscripts r and l respectively represent right and left. Both Z_{hip} and $\dot{Z}_{hip}$ were normalized to body height, bh. Using an existing sample of 53 young subjects, the ratio of hip height to bh is 51.0+ 1.4% ($R^2 = 0.82$, $p < 0.001$) in standing. To validate the use of hip motion to characterize limb support, the relationship between the impulse, $I(t)$, of the vertical component of the ground reaction force, $F_Z(t)$, and the resulting change in hip height, $\Delta Z_{hip}(t)$, was investigated. The vertical impulse and change in hip height were calculated as:

$$\begin{aligned} I(t) &= \int_{L-LO}^{t} \left[F_Z(t) - mg\right] dt \\ \Delta Z_{hip}(t) &= Z_{hip}(t) - Z_{hip,L-LO} \end{aligned}$$

Where, g and m are acceleration due to gravity and body mass, and L-LO represents the recovery foot liftoff. It was found that the vertical impulse was highly correlated with the changes in hip height at every instant during single-stance phase (Fig. 1), where $I(t) = 2.23\Delta Z_{hip}(t) - 0.008$ ($R^2 = 0.79$, $p < 0.001$). We therefore adopted simpler measurement of the two, i.e., the hip height.

The unexpected slip trial was analyzed at four gait-slip points-of-interest (Fig. 2-a): the slipping foot touchdown (R-TD) prior to slip onset, recovery foot liftoff (L-LO), the instant immediately before recovery touchdown (L-TD$_{pre}$) and its touchdown (L-TD$_{post}$). All events were determined from force plate data and verified using foot kinematics. The platform started moving approximately 24 ms after R-TD. The events in sit-to-stand-slip included seat-off (SO), the recovery liftoff (L-LO), the instant immediately prior to the recovery touchdown (L-TD_{pre}) and its touchdown (L-TD_{post}) (Fig. 2-b). The platform started moving approximately 16 ms after SO.

To simplify stability calculation for the sake of reducing discontinuity, the posterior border of the BOS was marked by the slipping heel before L-TD_{pre}, and by the recovery heel only at L-TD_{post}. Hip vertical motion remained the same from L-TD_{pre} to L-TD_{post}, and was designated only as L-TD. Recovery step length was obtained by subtracting recovery heel from slipping heel at TD_{post}. The stride length of the recovery limb was defined as the moving distance of the recovery heel from R-TD to L-TD_{post}. The slip distance was the travel displacement of slipping heel from R-TD to L-TD_{post}. The maximum slip distance was defined as the total movement that occurred along the tracks before the slipping heel stopped moving.

Slip outcomes were classified as falls, recoveries, and harness-affected based on the force recorded by the safety harness load cell (Table 1) (Brady et al., 2000). Fallers were confirmed via visual inspection of recorded video after the subjects were unambiguously supported by the harness (i.e., when load cell force >30% body weight as a cutoff criterion) and were identified as fallers. Harness-affected (i.e., when load cell force exceeded 4.5% body weight over any 1-second period after the slip occurs but did not exceed a peak of 30% body weight) and other unusable (due to technical reasons) trials were excluded (Table 1).

Two-way ANOVA with Bonferroni-corrected post hoc *t*-tests, were used to identify the differences between the slip outcomes (fall vs. recovery) and between the tasks (gait-slip vs. sit-to-stand-slip) in stability, BOS kinematics, hip motion state at all events, and the duration of bipedal and single-stance phases. The prediction accuracy of slip outcome was computed for both tasks, individually, using logistic regression with COM stability, limb support, and their combination as independent variables. Similar analysis was also performed with the task (i.e. the sit-to-stand-slip or gait-slip) as the covariate, to identify if it had a significant impact on slip outcome prediction. A significance level of 0.05 was used throughout. Analyses were performed using SPSS 15.0 (Chicago, IL).

RESULTS

For gait-slip, no between-group differences were detectable in stability or limb support at R-TD (Table 2). The fallers (n = 9, Table 1) had a delayed initiation of the recovery step ($p <$ 0.001, Fig. 3) with a longer slip distance ($p < 0.05$, Table 3) and a higher slip velocity ($p <$ 0.001 at L-LO and $p < 0.01$ at L-TD_{pre}) than those who recovered (n = 44, Table 1). They were less stable than those who recovered early on at L-LO, while stability continued to deteriorate during single-stance phase to its lowest point at L-TD_{pre} ($p < 0.001$, Table 2). Both outcome groups took a similar recovery step with the recovery step length, the stride length, and the duration of the step revealing no outcome-related differences (Table 3 and Fig. 3). Due to the change of BOS to the recovery limb from slipping limb, COM stability was positive at L-TD_{post}, and there was no between-group difference in stability (Table 2). Fallers demonstrated a lower Z_{hip} ($p < 0.001$, Table 2) and faster downward velocity $\dot{Z}_{hip}$ ($p < 0.001$, Table 2) than those who recovered early on at L-LO. This became more severe during single-stance phase, when fallers exhibited more rapid hip descent with a lower Z_{hip} than did the recovery group at L-TD ($p < 0.001$, Table 2). The results for the multi-step fallers (n = 4, Tables 1 and 2) were not analyzed due to small sample size.

No between-group differences were detected in stability or limb support at SO in sit-to-stand-slip (Table 2). In contrast to gait-slip, stability was not different at L-LO, but was significantly lower among fallers (n = 9, Table 1) than those who recovered (n = 32, Table 1) at both L-

TD_{pre} ($p < 0.05$) and L-TD_{post} ($p < 0.001$) (Table 2). Z_{hip} was lower in sit-to-stand-slip than it was in gait-slip at SO (R-TD for gait-slip) and at L-TD among fallers (Table 2). Another major between-task difference was that recovery step initiation was twice longer in sit-to-stand-slip as in gait-slip ($p < 0.001$, Fig. 3). There were no between-task nor between-outcome differences in the duration of single-stance phase ($p > 0.05$, Fig. 3).

In both tasks, stability and limb support played an important role in resisting falls. Jointly, both predicted 88.9% of gait-slip and 100% sit-to-stand-slip falls at L-TD_{pre} (Table 4). The predictability for both factors combined was reduced at L-TD_{post} (Table 4). Specifically, they can predict 55.6% and 88.9% of falls respectively for gait-slip and sitto-stand-slip at L-TD_{post} (Table 4). Stability could predict more falls at L-TD_{pre}, but less falls at L-TD_{post} in gait-slip than in sit-to-stand-slip (Table 4). At L-TD_{post}, limb support became the dominant predictor of slip outcome, as indicated by its higher prediction accuracy of falls than stability for both tasks (Table 4).

DISCUSSION

As postulated, the reactive control of stability and limb support together play a dominant role in resisting a slip-related fall regardless of the task. The slip severely destabilized all subjects, such that immediately prior to recovery step touchdown, the stability deteriorated to its lowest level among both fallers and those who recovered in both tasks (Table 2). Slip-induced instability is the precursor to a fall, whereby a recovery step must be taken following backward balance loss to avert an actual fall in response to an unannounced slip. Without that step, each backward balance loss could have resulted in a fall (Pavol and Pai, 2007). Yet, regaining stability alone is insufficient to prevent a fall. Because the recovery steps were highly effective in restoring stability among these subjects regardless of outcome and task, the predictive strength of stability diminished in gait-slip or reduced in sit-to-stand-slip after touchdown, while limb support remained able to differentiate fallers from those who recovered in both tasks. When slip-induced instability was combined with inadequate limb support, falls became nearly inevitable (with nearly 90% and 100% certainty in gait-slip and in sit-to-stand-slip, respectively).

As postulated, such insights on reactive control of stability and limb support may not be gained in this case prior to perturbation onset (Table 2). There were no detected differences between the fallers and those who recovered in any of the variables investigated during regular performance of both tasks prior to the onset of the novel, unannounced slip (i.e. R-TD for gait-slip and SO for sit-to-stand-slip). Hence, such findings could, in principle, raise the question of the evaluation, commonly based on the performance during volitional, *unperturbed* movements, to predict one's vulnerability to future falls (Berg et al., 1992;Tinetti, 1986). Reaction-based tests that mimic real-life situations where falls occur might provide better prognostic value in comparison to volitional-based tests, at least among health individuals, about the predictions on such risk.

Although at the onset of the slip there were no differences detectable in both stability and limb support between fallers and those who recovered, significant differences had already clearly appeared at recovery step liftoff for gait-slip (Table 2). The importance of this transitional phase from bipedal to single stance has been similarly noted in previous studies examining responses to forward slips (Lockhart et al., 2003;Tang and Woollacott, 1998;You et al., 2001). During this period, the body weight is transferred from the contralateral limb to the slipping limb, while unintended vertical descent may be initiated. After recovery step liftoff, real-time feedback adjustments might not be sufficient to alter the motor program of forward stepping, due to the short duration of the single stance phase (Tripp et al., 2004). Most prominently, in spite of the differences in the outcomes and in the tasks, the fact that step execution time (and step length across different outcomes) remained invariant suggests the commonality existing in motor programming pertaining to its temporal (and spatial) characteristics of this triggered response.

There were noticeable task-specific differences in stability, which may result from differences in the control mechanisms that influence the recovery following a slip. The greater forward COM momentum during gait-slip than sit-to-stand-slip results in doubling the BOS velocity during gait-slip compared to sit-to-stand-slip (Table 3). Such increased slip intensity contributes to a lower stability at recovery step liftoff in gait-slip and predisposes individuals to a fall. This advantage can later be compromised by a prominent task-specific delay, doubling the response time in initiating the recovery step (Fig. 3). Although recovery steps against backward fall should all land posterior to the slipping limb, a more rapid step response could be attributed to the possibility that forward stepping is part of the regular motor program for gait but not for sit-to-stand. Presumably, subjects in gait-slip would only need to *modify* their ongoing motor program rather than *initiate* a new stepping program, as in sit-to-stand-slip. Moreover, for sit-to-stand, length of the *backward* step could be influenced by greater anatomical and physiological constraints, adding another level of difficulty resulting in only half of the step length in comparison to *forward* step in gait. Paradoxically, a shorter forward recovery step would be more desirable in gait-slip, and a longer step length only reveals the limitation in motor program modification.

The reactive control of stability is directly reflected in the control of the slip kinematics. The relative motion between COM and BOS is highly associated with slip recovery outcome (You et al., 2001). Our averaged maximum slip distance was as high as 0.78m with peak slip velocity of 2.51m/s (Table 3), which appeared to be higher than previously reported. For example, the slip distance and peak velocity reported were at the levels of >0.1m and 0.78m/s (Redfern et al., 2001), >0.1m and ~1.5m/s (Moyer et al., 2006), 0.1m and 0.5m/s (Strandberg and Lanshammar, 1981), ~0.1m and 1.17m/s (Lockhart and Kim, 2006), and 0.34m and 1.58m/s (Troy and Grabiner, 2006). Despite such differences, our results confirmed previous findings that slip distance and slip velocity potentially associated with falls (Beschorner and Cham, 2008; Cham and Redfern, 2002; Lockhart and Kim, 2006; Redfern et al., 2001; Strandberg and Lanshammar, 1981). The BOS velocity and its position directly affect the COM stability (Yang and Pai, 2009), and the reactive control of stability and limb support does indeed differentiate the fallers from those who recovered in gait-slip, especially prior to step touchdown (Table 4).

Executing the recovery step significantly improved stability for both fallers and those who recovered in both tasks. A successful recovery step diminished the differences in stability between fallers and recoveries; leaving limb support to play the dominant role after step touchdown (Tables 2 and 4). Nonetheless, there might be some tradeoff between providing limb support and regaining stability. The change from bipedal to single stance might reduce the amount of vertical limb support being provided during the crucial time period, when the slipping limb's hip descent has already begun and continues to rapidly deteriorate in both tasks. It was postulated that in sit-to-stand, the reduction in limb support due to the change from bipedal to single stance could initiate or further hasten hip descent, if the withdrawal of the recovery (stepping) limb was not adequately compensated by the stance (slipping) limb (Pai et al., 2006). From this perspective, the gains in stability through stepping could have been made at the expense of limb support upon an unexpected slip. The rapid hip descent consistently found in both fallers and those who recovered from liftoff to touchdown in gait-slip further supports this notion (Fig. 2 and Table 2).

It has been reported that some gait factors, that can be evaluated prior to the initiation of a slip, such as heel orientation of the foot at heel contact, step length and cadence (Holbein-Jenny et al., 2007; Moyer et al., 2006), heel contact velocity (Lockhart and Kim, 2006), and heel contact acceleration (Beschorner and Cham, 2008) are related to slip-initiated falls. Similarly, in this study, unperturbed gait stability would tend to be lower in fallers, especially multi-step fallers (Table 2, R-TD). Nonetheless, the reactive response could in fact provide a much better assessment of one's vulnerability to falls upon a novel, unannounced slip than that of volitional performance assessed prior to the onset of slip. It is noteworthy; however, the findings of this and other studies are dependent upon the specific conditions under which slips are induced. We acknowledge that the differences in the slip-inducing mechanisms (Beschorner and Cham, 2008; Cham and Redfern, 2001; Holbein-Jenny et al., 2007; Lockhart et al., 2003; Moyer et al., 2006), in the aforementioned slip intensity, and even in sample population (e.g., young versus old, healthy versus frail) or in sample size may all affect such findings. Further, a person's reaction to unexpected perturbation with recovery stepping would logically be very different from his/her regular gait pattern, and hence they each could reveal different insights about this person's ability.

The appropriateness of using hip height to characterize limb support has been investigated in the present study. Alternatively, the impulse resulting from vertical ground reaction force could be applied to characterize the limb support instead of hip height. The relationship between change in hip height and impulse resulted from vertical ground reaction force as well as body gravity has been investigated and a high linear relationship between them was found (Fig. 1). Given such high linear correlation, it is reasonable to anticipate that both characterizations of limb support lead to the same conclusion (please see Methods), hip height is preferred because it provides a variable quantifiable in clinics without the need of force platforms. It is also noteworthy that the multi-step fallers were not included in the analysis due to their small sample size. If only 4 of 67 young adults as in the present study exhibit multi-step falls, it would require 201 subjects to detect significant differences in stability between recoveries and multi-step fallers at 80% power with effect size of 1.05 in gait-slip. This exclusion has unfortunately reduced the overall sample size of the fallers that could be used in the analyses. Nonetheless, this unintended consequence has indeed revealed that majority of the falls took place within *one step* regardless of the task.

In conclusion, averting an actual fall following an unannounced, novel slip requires the ability to restore stability by taking a successful recovery step while providing sufficient limb support to retard unintended hip descent. Our results suggest that reactive control of these two factors governs the recovery outcome in both activities, more accurately than during the unperturbed part of the movement. Future assessment may need to properly include and evaluate *both* predictors in order to accurately assess a person's reaction and risk of falls, while the selection of one activity versus the other maybe relatively less critical.

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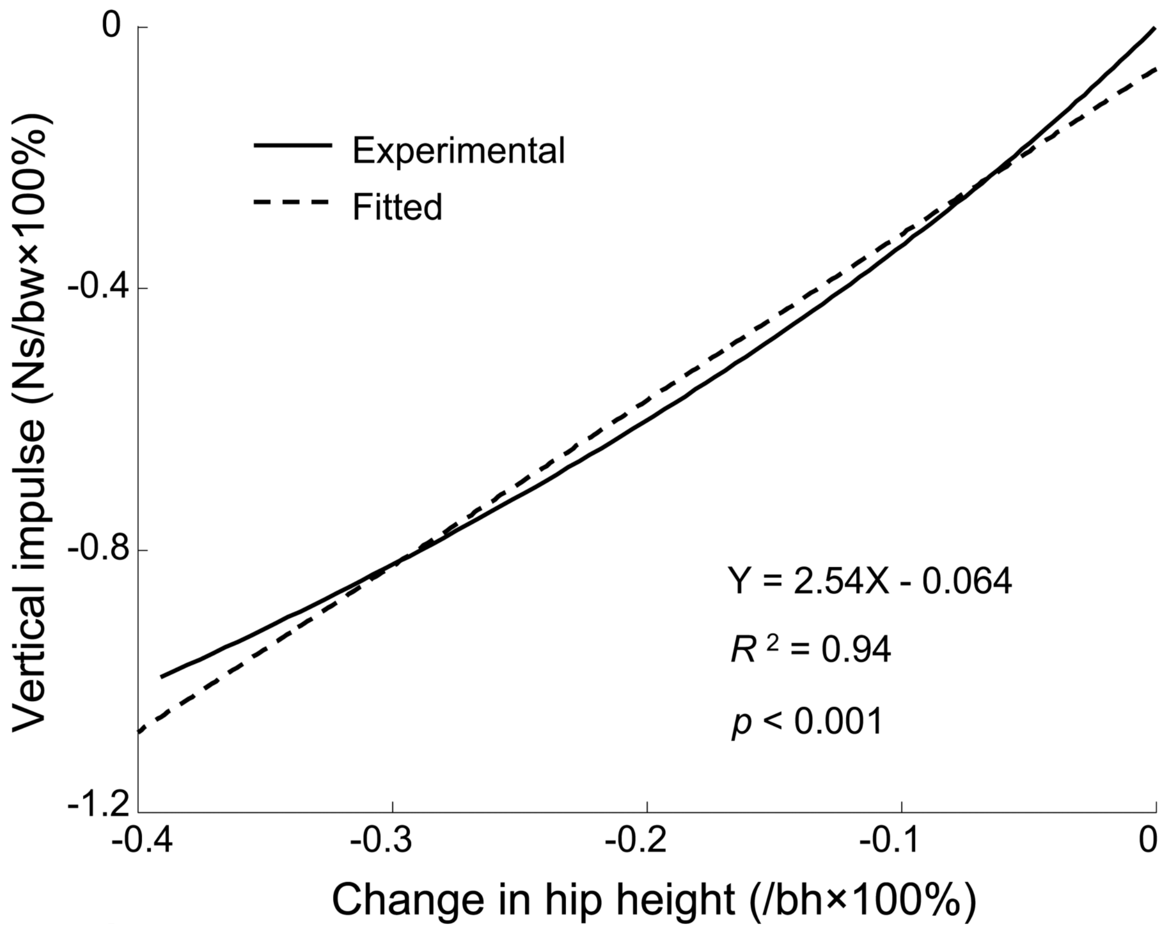

Fig.1.

A typical plot of the linear relationship between impulses resulted from vertical component of the ground reaction force (F_Z) as well as body gravity and the resulting change in hip height during single stance phase from recovery liftoff (L-LO) to its touchdown for a gait-slip fall trial. The experimental results (solid line) are well fitted by a straight line (dashed line). The impulse is calculated as $\int_{t_{L-LO}}^{t} [F_Z(t) - bw]\, dt$, where *bw* represents body weight. The change in hip height is with respect to hip height at L-LO. The vertical impulse is normalized to *bw*, and change in hip height normalized to body height, *bh*.

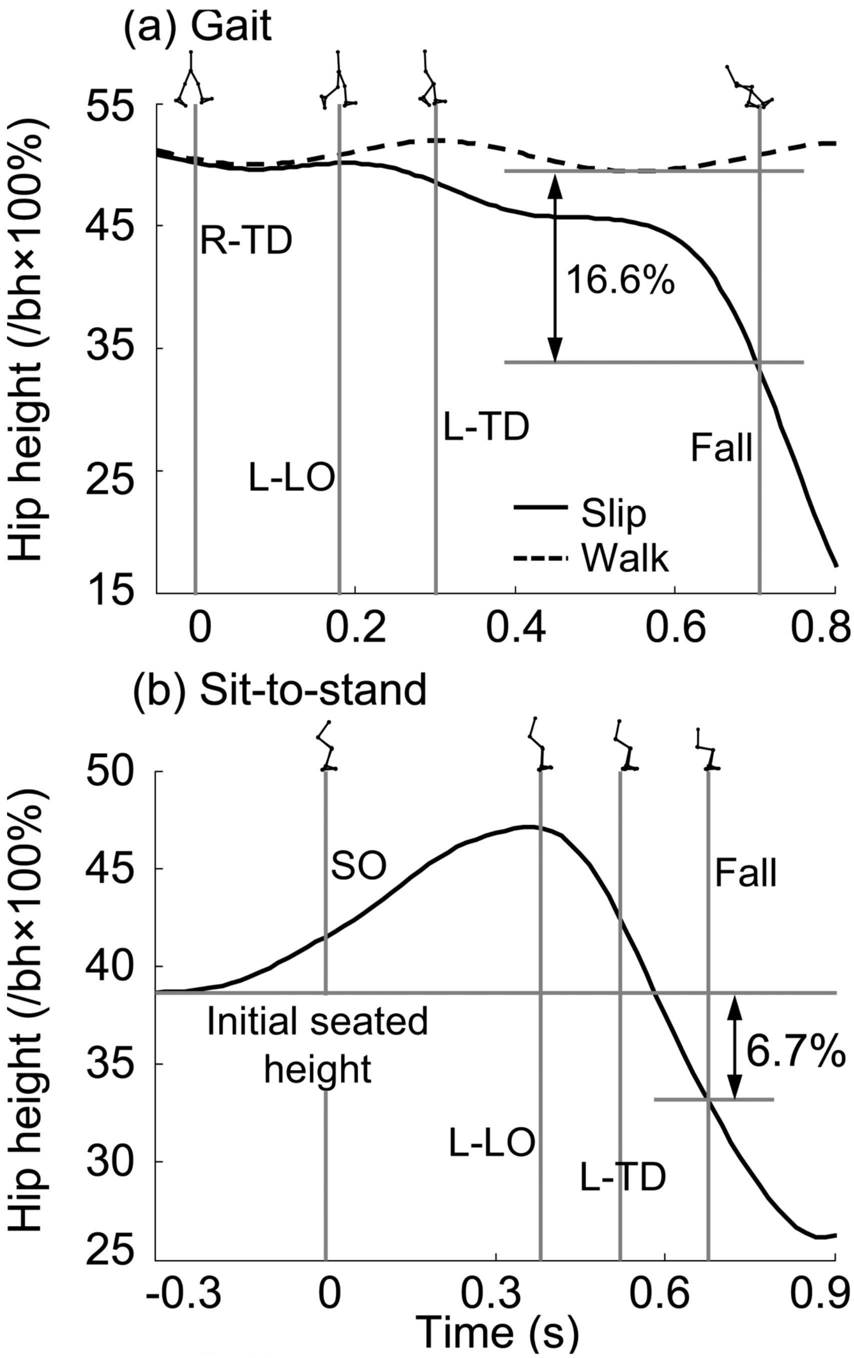

Fig. 2.

Typical plots of the time history of mid-hip point height during (a) gait-slip and (b) sit-to-stand-slip. Also shown are the events of interest (vertical lines). They include slipping (right) limb touchdown (R-TD) in gait and seat-off (SO) in sit-to-stand, recovery step liftoff (L-LO) and touchdown (L-TD). The actual fall time, which is identified as the instant when the fall-arrest force exerted on the load cell exceeds 30% body weight, is also marked as a vertical line. Horizontal lines illustrate the values of hip descent at the actual fall time.

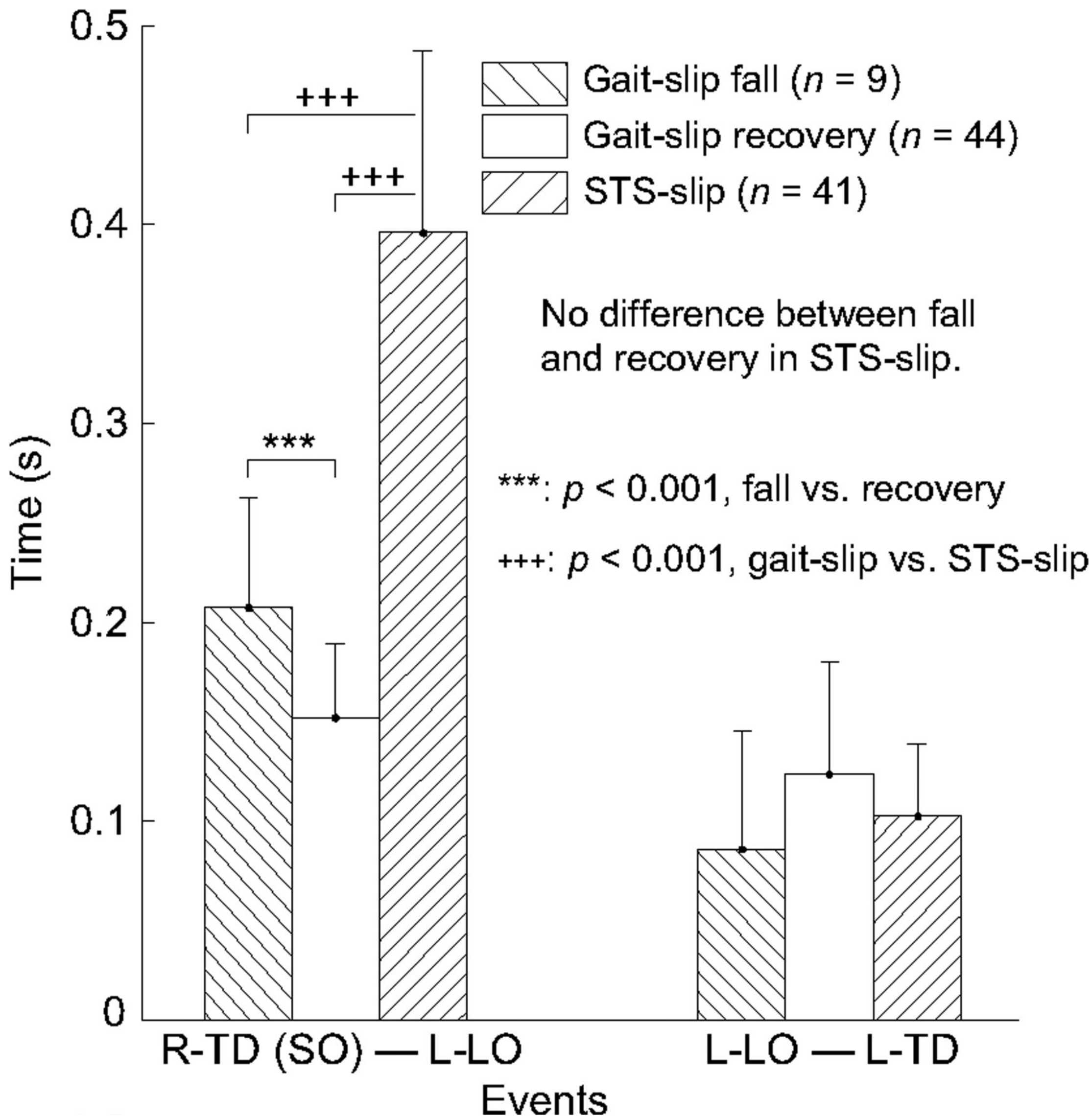

Fig. 3.

Group mean (column height) and standard deviation (bar) of the elapsed time of bipedal phase from slipping foot touchdown (R-TD) for gait-slip or seat-off (SO) for sitto-stand-slip to recovery liftoff (L-LO), and single-stance phase from L-LO to recovery touchdown (L-TD) for gait-slip fall ($n = 9$), gait-slip recovery ($n = 44$), and sit-to-stand-slip (STS, $n = 41$) groups. Because there is no difference between fallers and those who recovered in both durations for STS-slip, the values from fall and recovery groups are combined. The time consumed by the bipedal phase in our study was comparable to previously reported results (You et al., 2001).

Table 1

Comparisons of experimental setup and analysis between gait-slip and sit-to-stand-slip tasks

		Gait-slip	Sit-to-stand-slip
Tasks		After ~10 walking trials, the 1st unannounced slip	After ~4 regular trials, the 1st unannounced slip
Release mechanisms		Pre-released movable platform prior to foot contact	Both movable platforms were released at seat-off
Maximum allowed slip distance		1.50 m	0.24 m
Slip outcome classification criterion	Fall	Peak harness force ≥ 30% *bw*†	Peak harness force ≥ 30% *bw*
	Recovery	Average harness force < 4.5% *bw* over any 1-second period	Average harness force < 4.5% *bw* over any 1-second period
	Harness-affected	None of above	None of above
Slip outcome (total *n* = 67 for gait; and *n* = 60 for sit-to-stand)	Single-step fall	9 or [9/(9+4+44)=16%]	9 or [9/(9+3+32)=20%]
	Multi-step fall	4 or [4/(9+4+44)=7%]	3 or [3/(9+3+32)=7%]
	Recovery	44 or [44/(9+4+44)=77%]	32 or [3/(9+3+32)=73%]
	Harness-affected	5	6
	Excluded*	5	10

†body weight

*the exclusive criteria include missing data or trigging problem..

Table 2

Mean (SD) kinematics of COM stability and of limb support as a function of slip outcome for gait-slip and sit-to-stand-slip (STS) at investigated events, which included slipping (right) limb touchdown (R-TD) in gait and seat-off (SO) in sit-to-stand to characterized preslip behavior. Recovery foot liftoff (L-LO), immediate pre recovery foot touchdown ($L\text{-}TD_{pre}$) and its touchdown ($L\text{-}TD_{post}$) are also analyzed for both tasks. Hip motion remains unchanged between the latter two.

	Stability (× 10^{-2})			**Hip height (/bh × 100%)**			**Hip velocity (/bh/s × 100%)**		
Gait	Single-Step Fall (*n* = 9)	Recovery (*n* = 44)	Multi-Step Fall (*n* = 4)	Single-Step Fall	Recovery	Multi-Step Fall	Single-Step Fall	recovery	Multi-step fall
R-TD	−14.4(4.9)	−12.9(4.6)	−16.3(4.6)	48.1(1.4)	48.3(1.5)	46.7(2.7)	−5.9(6.7)	−2.8(6.2)	−3.9(1.6)
L-LO	−35.8(8.8)^{b}	−19.8(9.7)	−28.9(13.4)	48.3(1.3)^{b}	49.9(1.7)	47.6(2.7)	−4.8(8.6)^{b}	3.3(7.1)	−0.4(0.5)
$L\text{-}TD_{pre}$	−47.4(11.2)^{c}	−29.4.0(12.0)	−37.0 (14.7)	47.1(1.5)^{c}	49.1(1.6)	47.1(2.3)	−25.2(5.9)^{c}	−13.3(8.4)	−9.9(9.2)
$L\text{-}TD_{post}$	92.7(26.2)	88.9(19.6)	91.5(25.5)						
STS	Single-Step Fall (*n* = 9)	Recovery (*n* = 32)	Multi-Step Fall (*n* = 3)	Single-Step Fall	Recovery	Multi-Step Fall	Single-Step Fall	recovery	Multi-step fall
SO	5.5(2.1)^{a}	7.2(3.9)^{a}	3.6(4.2)	41.8(2.0)^{a}	41.9(1.6)^{a}	42.8(2.2)	36.5(8.1)^{a}	37.1(8.4)^{a}	33.4(7.0)
L-LO	−11.3(7.7)^{a}	−8.9(10.4)^{a}	−31.1(10.5)	49.5(1.3)	49.6(1.8)	49.4(2.0)	−30.5(20.0)^{a}	−31.9(3.3)^{a}	−8.6(10.1)
$L\text{-}TD_{pre}$	−32.0(10.8)^{a,d}	−20.6(13.9)^{a}	−33.7(9.1)	43.6(1.6)^{c,e}	49.5(1.6)	47.6(3.1)	−40.2(14.7)^{a,c}	−29.7(9.8)^{a}	−12.4(9.3)
L_TD_{post}	5.5(15.0)^{a,f}	22.5(9.8)^{a}	6.9(13.9)						

Statistical analysis was not performed for multi-step fallers due to small sample size.

[a] $p < 0.001$ vs. gait-slip (main effect of task)

[b] $p < 0.001$ vs. gait-slip recovery (interaction of task and slip outcome)

[c] $p < 0.001$ vs. recovery (main effect of slip outcome)

[d] $p < 0.05$ vs. recovery (main effect of slip outcome)

[e] $p < 0.05$ vs. gait-slip fall (interaction of task and slip outcome)

[f] $p < 0.001$ vs. STS-slip recovery (interaction effect of task and slip outcome).

Table 3

Group mean ± standard deviation of recovery step length (X_{step}) at recovery foot touchdown, the stride length X_{stride} of recovery limb, slip distance (X_{slip}) and velocity of base of support (BOS) ($V_{BOS, L\text{-}TD, pre}$) at the instant immediately prior to recovery touchdown, maximum slip distance ($X_{slip, max}$), and peak slip velocity of BOS, ($V_{BOS, max}$) for fall and recovery, and for slip and non slip (unperturbed) trials during both gait and sit-to-stand (STS) tasks.

Tasks	Variables	Slip		Non Slip
Gait		Fall (*n* = 9)	Recovery (*n* = 44)	(*n* = 53)
	X_{step} (m)	−0.446 ± 0.122	−0.520 ± 0.134	0.646 ± 0.074
	X_{stride} (m)	0.427 ± 0.190	0.418 ± 0.175	1.301 ± 0.127
	X_{slip} (m)	0.281 ± 0.066[a]	0.221 ± 0.066	0.015 ± 0.009
	$V_{BOS, L\text{-}TD, pre}$ (m/s)	1.750 ± 0.270[b]	1.391 ± 0.340	0.001 ± 0.004
	$X_{slip, max}$ (m)	0.781 ± 0.333[b]	0.560 ± 0.293	0.017 ± 0.009
	$V_{BOS, max}$ (m/s)	2.513 ± 0.280[c]	1.982 ± 0.449	0.001 ± 0.007
STS		Fall (*n* = 9)	Recovery (*n* = 32)	(*n* = 41)
	X_{step} (m)	−0.207 ± 0.112[e]	−0.243 ± 0.099[e]	
	X_{stride} (m)	−0.127 ± 0.100[e]	−0.160 ± 0.103[e]	
	X_{slip} (m)	0.256 ± 0.008[a]	0.240 ± 0.037[Δ]	0.011 ± 0.006
	$V_{BOS, L\text{-}TD, pre}$ (m/s)	−0.038 ± 0.107[d,e]	0.367 ± 0.580[e]	0.001 ± 0.003
	$X_{slip, max}$ (m)	0.258 ± 0.008[e]	0.260 ± 0.012[Δe]	0.013 ± 0.007
	$V_{BOS, max}$ (m/s)	1.305 ± 0.256[e]	1.331 ± 0.304[e]	0.001 ± 0.004

[a] $p < 0.05$ vs. recovery (main effect of slip outcome)

[b] $p < 0.01$ vs. gait-slip recovery (interaction of task and slip outcome)

[c] $p < 0.001$ vs. gait-slip recovery (interaction of task and slip outcome)

[d] $p < 0.001$ vs. STS-slip recovery (interaction of task and slip outcome)

[e] $p < 0.001$ vs. gait-slip (main effect of task).

[Δ] Travel displacement of slider may exceed 0.24 m, which was the design specification, due to the deformation of the shock absorber from the impact during an actual slip trial.

Table 4

Comparison of the prediction accuracy (%) of the fall incidences based on the center of mass (COM) stability, the limb support, and the combination of the COM stability and limb support at the investigated events in gait-slip and in sit-to-stand-slip. The events included slipping (right) limb touchdown (R-TD) in gait and seat-off (SO) in sit-to-stand to characterized preslip behavior. Recovery foot liftoff (L-LO), immediate pre recovery foot touchdown ($L\text{-}TD_{pre}$) and its touchdown ($L\text{-}TD_{post}$), (or simply L-TD for hip height measurement) are also analyzed for both tasks.

	Event	Gait (*n* = 9)	Sit-to-stand (*n* = 9)	Task-Specific
Stability	R-TD (SO)	0	0	$N^{\#}$
	L-LO	44.4	0	Y^{Δ}
	$L\text{-}TD_{pre}$	44.4	11.1	Y^{Δ}
	$L\text{-}TD_{post}$	0	55.6	Y^{Δ}
Limb support	R-TD (SO)	0	0	$N^{\#}$
	L-LO	33.3	11.1	Y^{Δ}
	L-TD	66.7	77.8	$N^{\#}$
Stability and limb support	R-TD	0	0	$N^{\#}$
	L-LO	66.7	11.1	Y^{Δ}
	$L\text{-}TD_{pre}$	88.9	100	$N^{\#}$
	$L\text{-}TD_{post}$	55.6	88.9	$N^{\#}$

Y^{Δ}: $p < 0.05$, the prediction accuracy is task-specific.

$N^{\#}$: $p > 0.05$, the prediction accuracy is not task-specific.