Role of Feedforward Control of Movement Stability in Reducing Slip-Related Balance Loss and Falls Among Older Adults

Y.-C. Pai, J. D. Wening, E. F. Runtz, K. Iqbal, and M. J. Pavol

INTRODUCTION

A significant health threat facing the older adult population is their increasing susceptibility to falling with increasing age (Baker and Harvey 1985; Holbrook et al. 1984; Tinetti et al. 1988). Falls are the leading cause of injury-related death or hospitalization in this older population (Baker and Harvey 1985). If new and effective clinical training strategies to reduce the risks of falls in the elderly are to be devised, one important requirement is an understanding of the mechanisms whereby the CNS regulates movement and stability (Maki and McIlroy 1996; Stelmach and Worthingham 1985; Woollacott et al. 1999).

Because human upright posture is inherently unstable, a primary objective for the CNS must be to prevent falls, achieved first by preventing unintended loss of balance. Loss of balance occurs when the motion state (i.e., instantaneous position and velocity) of the body center-of-mass (COM) with respect to the base of support (BOS) exceeds certain stability limits (Maki 1998; Pai 2003). It is possible that the CNS can integrate afferent inputs of different origins to monitor and update the current COM state and readily compare it with a corresponding internal representation of these stability limits. Adaptive refinement of the internal representation of postural stability to account for real or potential perturbation may be required to improve the CNS’s ability to prevent balance loss. The CNS can then select and execute an appropriate action in a feedforward control manner, to counter the perturbation and to avert any unintended balance loss.

It is logical to postulate that, relying on prior experience and memory, the CNS must be able to quantify the likelihood of balance loss. This ability would likely require the mapping of stability limits, possibly in terms of a feasible stability region in the COM state space (Pai and Patton 1997). Outside of this region, the tasks of movement termination and balance recovery for upright standing can never be simultaneously successful. Theoretically, these stability limits can also be deduced mathematically based on assumed stability criteria, the dynamics of the body, anatomical and physiological limitations, and environmental constraints. For its verification and the demonstration of the potential of its practical application, this concept can be applied to theorize the prevention of slip-related falls. It is predicted that a backward balance loss can be avoided through the use of feedforward control to improve stability at the onset of a slip. Specifically, one can increase the forward COM velocity and/or anteriorly shift the COM to achieve this objective (Pai and Iqbal 1999).

The same mathematical model simulation predicts the existence of a set of “optimal” movement strategies that satisfy the constraints associated with avoiding a loss of balance under the hypothesis that an adaptive improvement in feedforward control relies on an accurate internal representation of stability limits, which must be a function of anatomical, physiological, and environmental constraints and thus should be computationally deducible based on physical laws of motion. We combined an empirical approach with mathematical modeling to verify the hypothesis that an adaptive improvement in feedforward control of COM stability correlated with a subsequent reduction in balance loss. Forty-one older adults experienced a slip during a sit-to-stand task in a block of slip trials, followed by a block of nonslip trials and a re-slip trial. Their feedforward control of COM stability was quantified as the shortest distance between its state measured at seat-off (slip onset) and the mathematically predicted feasible stability region boundary. With adaptation to repeated slips, older adults were able to exponentially reduce their incidence of falls and backward balance loss, attributable significantly to their improvement in feedforward control of stability. With exposure to slip and nonslip conditions, subjects began to select “optimal” movements that improved stability under both conditions, reducing the reliance on prior knowledge of forthcoming perturbations. These results can be fully accounted for when we assume that an internal representation of the COM stability limits guides the adaptive improvements in the feedforward control of stability.

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Pai, Y.-C., J. D. Wening, E. F. Runtz, K. Iqbal, and M. J. Pavol. Role of feedforward control of movement stability in reducing slip-related balance loss and falls among older adults. J Neurophysiol 90: 755–762, 2003; 10.1152/jn.01118.2002. Human upright posture is inherently unstable. To counter the mechanical effect of a large-scale perturbation such as a slip, the CNS can make adaptive adjustments in advance to improve the stability of the body center-of-mass (COM) state (i.e., its velocity and position). Such feedforward control relies on an accurate internal representation of stability limits, which must be a function of anatomical, physiological, and environmental constraints and thus should be computationally deducible based on physical laws of motion. We combined an empirical approach with mathematical modeling to verify the hypothesis that an adaptive improvement in feedforward control of COM stability correlated with a subsequent reduction in balance loss. Forty-one older adults experienced a slip during a sit-to-stand task in a block of slip trials, followed by a block of nonslip trials and a re-slip trial. Their feedforward control of COM stability was quantified as the shortest distance between its state measured at seat-off (slip onset) and the mathematically predicted feasible stability region boundary. With adaptation to repeated slips, older adults were able to exponentially reduce their incidence of falls and backward balance loss, attributable significantly to their improvement in feedforward control of stability. With exposure to slip and nonslip conditions, subjects began to select “optimal” movements that improved stability under both conditions, reducing the reliance on prior knowledge of forthcoming perturbations. These results can be fully accounted for when we assume that an internal representation of the COM stability limits guides the adaptive improvements in the feedforward control of stability.

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both slip and nonslip conditions (Pai and Iqbal 1999). Such movement options are optimal because they simultaneously reduce the likelihood of a balance loss under both possible conditions when facing the uncertainty that a slip may or may not occur. Thus they can lessen the reliance of the CNS on detailed and accurate knowledge of a forthcoming balance perturbation.

Recent empirical evidence showed that fall incidence in older adults decreased with repeated exposure to slipping and nonslipping conditions (Pavol et al. 2002b) and that this decrease was associated with anticipatory (proactive) adjustments to COM state (Pavol et al. 2002d). It is still unclear, however, the extent to which stability at slip onset, as quantified through the feasible stability region concept, can actually explain reductions in backward balance loss and fall incidence. Existence of such a relationship would lend support to the feasible stability region concept as a conceptual model of the hypothesized CNS internal representation of stability limits, thereby supporting its application to fall prevention. Further, this relationship would support the theory that the internal representation of stability limits can be rapidly refined (i.e., updated or modified) through repeated perturbation exposure.

The purpose of this study was to verify this concept of adaptive feedforward control of movement (dynamic) stability by testing three specific hypotheses. First, movement stability can be improved among older adults through repeated slip exposure, such that the improvement correlates with a reduction in the likelihood of backward balance loss that, in turn, should be associated with a reduction in fall incidence. Conversely, a reduction in movement stability against forward balance loss due to overcompensation will correlate with an elevated risk of forward balance loss, which can again be reduced with an adaptive improvement in movement stability. Last, the predicted optimal movement strategies to counter the uncertainty of a slip are attainable, such that movement stability is achieved under both slip and nonslip conditions and the likelihood of both forward and backward balance loss is reduced.

FIG. 1. A: free body diagram of the 2-link (feet + rest-of-body) inverted pendulum model of the human body (left), and definition of variables (right). B: the equations of motion, where \( m \) is the mass of the body minus feet; \( m_f \), the mass of the feet; \( r \), the length to the pendulum mass center; \( \theta \), the angular position of the body segment; \( x \), the displacement of the base of support; \( \tau \), the ankle joint moment; \( F_x \), the horizontal component of the ground reaction force; \( F_y \), the vertical component of the ground reaction force acting at the center of pressure (COP); \( \mu \), the coefficient of friction; \( g \), the acceleration due to gravity; and \( c \): a schematic representation of the model simulation and optimization process.

METHODS

Mathematical derivation of feasible stability region

The feasible stability region is defined as all combinations of COM anteroposterior and velocity for which a loss of balance is preventable. Loss of balance for a given initial position and velocity occurs when this velocity of the COM relative to the BOS cannot be reduced to zero within the existing BOS limits, but only with a change in the BOS. To search for the boundaries of the feasible stability region, we used a two-link model (Fig. 1a) with an optimization control loop (Fig. 1c). One model segment represented the symmetrical placement of the feet and the second segment represented the rest of the body. The equations of motion for this two-link model with two degrees-of-freedom under slipping conditions (Pai and Iqbal 1999) are listed in Fig. 1a.

Forward dynamic solutions for these equations were derived by numerical integration using a fourth-order Runge-Kutta method, where initial conditions were the initial body state and joint moment estimates. The model was controlled through joint moments, which were parameterized as a mathematical function exhibiting sigmoid variation with time. The outputs of the simulation included time-histories of the horizontal and vertical components of the ground reaction force and the COM position and velocity (Pai and Iqbal 1999).

Optimization entailed an iterative process of movement simulation, evaluation of the cost function from the simulation results, and updating the model inputs based on the method of steepest descent (Pai and Iqbal 1999). The task objectives of a successful movement termination were quantified through a cost function. It incorporated mathematical expressions representing the desired final state of the model, the anatomical (e.g., joint range of motion) and physiological (e.g., muscle strength) limitations, the environmental constraints (e.g., characteristics of the ground reaction force), and the limits on the parameters that defined the joint moment profiles. The maximum horizontal ground reaction force component was determined by the coefficient of friction.

The solution derived from the simulation and optimization process determined, for a given COM position, the minimum initial COM velocity at which a backward loss of balance could be avoided. This process was repeated at other COM positions and for forward loss of balance. Polynomial interpolation between solutions was used to outline the boundary of the feasible stability region. Separate feasible
stability regions were determined for slipping and nonslipping conditions, using the corresponding coefficients of friction.

**Subjects**

Following approval by the Institutional Review Board, 41 healthy older adults (21 women) gave written informed consent and were paid to participate. They were ambulatory, community-dwelling individuals ≥65 yr of age (mean ± SD age: 73 ± 5 yr; height: 1.69 ± 0.09 m; mass: 79 ± 14 kg). Subjects were screened for the following exclusionary factors: neurological, musculoskeletal, cardiopulmonary, and other systemic disorders, selected drug usage (e.g., tranquilizers), cognitive impairment, poor mobility, and orthostatic hypotension. Calcaneal bone mineral density was assessed and individuals with bone loss (i.e., osteopenic or osteoporotic) were excluded to reduce the risk of fracture on an actual, harness-affected fall.

**Experimental protocol and data collection**

Slips were induced during a sit-to-stand movement using a protocol that has been detailed previously (Pavol et al. 2002b). Trials began with subjects sitting on a stool in a standardized position such that the heels were aligned, knees flexed to 100° from the anatomic position, and ankles at 10° dorsiflexion. After four regular sit-to-stand trials, a block of five consecutive slip trials (trials S-1 through S-5) was introduced without warning. This was followed by a block of three nonslip trials (trials NS-1 through NS-3). Subjects were then exposed to another slip trial (the re-slip trial, RS-1). Subjects were originally informed that they would initially be performing sit-to-stand trials and that “later on” a slip would take place. No practice was given, and the exact trial, timing, and mechanisms of the slip were not provided. After the first slip, subjects were informed that a slip “may or may not occur” during subsequent trials.

Slips were induced using two low-friction platforms (dimensions: 31 × 29 cm, friction coefficient: 0.02) placed side-by-side such that each foot rested on its own platform. Slips were initiated by a computer-controlled release of the low-friction platforms when the weight on the seat fell below 10% of body weight as measured by a force plate (AMTI, Newton, MA). As a result of rapid unloading, coupled with sampling and mechanical delays, the load applied to the seat by the subject reached zero 0.013 ± 0.010 s prior to movement of the sliding platforms. On release, the platforms moved forward freely and independently. After a maximum travel of 24 cm, the platform locked in the forward position. At least one platform traveled the maximum distance in 99.2% of all slips by all subjects. The mean duration for the first slip was 0.43 ± 0.10 s, with the platform reaching a mean peak velocity and acceleration of 0.86 ± 0.17 m/s and 8.18 ± 2.24 m/s², respectively. Figure 2 shows a time history of the COM and BOS position and velocity, as well as the vertical ground reaction force for a representative slip. Subjects wore a full-body safety harness attached at the shoulders to a ceiling-mounted support by a pair of shock-absorbing dynamic ropes, typically used for fall protection in rock climbing. Rope lengths were adjusted so the knees could not touch the flooring. A load cell monitored the force exerted on the ropes.

The kinematics of markers attached to the bilateral upper and lower extremities, torso, and platforms were recorded by a motion capture system at 60 Hz (Peak Performance, Englewood, CO). Marker paths were low-pass-filtered at marker specific cut-off frequencies ranging from 4.5 to 9 Hz using a recursive, fourth-order Butterworth filter. Locations of joint centers, heels, and toes were computed from the marker paths. These data were used to determine the anterior position and forward velocity of the COM with respect to the rear of the BOS (i.e., the heel of the posterior foot in ground contact) for each trial based on anthropometric data (Pavol et al. 2002a). The COM position and velocity were expressed as dimensionless fractions of $l_{BOS}$ and $\sqrt{g \times h}$ (McMahon 1984), respectively, where $l_{BOS}$ is the length of the base of support, $g$ is the acceleration due to gravity, and $h$ is the body height.

![Figure 2](https://example.com/figure2.png)

**Analysis and statistics**

A fall was defined based on the vertical descent of the hips after slip onset, occurring if the midpoint between the bilateral hip joint centers descended below 5% body height above its initial seated height. Other trials were considered harness-affected if the average force on the ropes exceeded 4.5% body weight over any 1-s period. The remaining trials were considered recoveries. Classification thresholds were determined post-hoc from clear divisions in the data distributions and were confirmed by the inspection of video recording images. A balance loss was determined to have occurred if a subject stepped to regain balance, that is, took a step that extended the BOS in the direction of stepping. The direction of the first such step was considered the direction of balance loss. If a subject recovered and did not step to regain balance, no balance loss occurred. For fall or harness-affected trials in which the subject did not step to regain balance, the direction of balance loss was determined from the position of the COM at the defined time of fall or start of harness effects, respectively. A COM position anterior to the more anterior toe or posterior to the more posterior heel corresponded to a forward or backward balance loss, respectively. Occasionally, due to equipment malfunction or experimenter error data were lost or a slip did not occur as intended. Six such trials were excluded from analysis.

The stability of a movement could be assessed by comparing the corresponding COM state trajectory to the mathematically derived feasible stability region boundaries for forward or backward balance loss under slip or nonslip conditions (Fig. 3). At seat-off, the stability against backward balance loss under slip conditions was quantified as the shortest distance between the instantaneous COM state and the
boundary for backward balance loss under slip conditions ($d$ in Fig. 4). Because the risks of forward and backward balance loss exist simultaneously, when expressed as a fraction of the corresponding width of the feasible stability region this measure also quantifies the stability against forward balance loss (Fig. 4). The stability under nonslip conditions at seat-off was computed similarly, based on the corresponding feasible stability region for nonslip conditions.

Our model predicts that, based on anatomical and physiological limitations and environmental constraints, a backward loss of balance must occur for COM states outside the corresponding boundary of the feasible stability region. Thus values $<0$ or $>1$ correspond to a predicted backward or forward balance loss, respectively. Backward balance loss should not occur when the stability measure is above the predicted threshold for backward balance loss ($0 \leq d < b$), because the COM forward momentum is sufficient to carry the COM forward from its current position to catch up with the BOS if the motor response (i.e., joint moments) is appropriate. Our rationale is that the farther inside the feasible stability region and away from the backward balance loss boundary the COM state is at slip onset, the greater the allowable deviations in the subsequent motor response, hence the greater the likelihood of avoiding a backward loss of balance through the response employed. Greater values of the above-defined stability measure are therefore taken to reflect greater stability against subsequent backward balance loss. An identical rationale is applicable to forward balance loss, although the numerical relationship between stability and the convention of the defined measure becomes both reversed and centered at 1 instead of 0. Values of the stability measure below the threshold for forward balance loss ($d < b$) reflect an increase in stability relative to forward loss of balance. A COM stability at slip onset that is $>1$ will be very favorable for avoiding a backward loss of balance, but will unavoidably result in a forward loss of balance.

An adaptive effect across trials of repeated slip exposure on reducing the incidence of backward balance loss and falls (vs. recoveries) was tested for the slipping block using a nonlinear (exponential) regression model. Logistic regression analyses determined the relationships between the mathematically predicted stability at seat-off and the corresponding probability of balance loss under the same conditions. Data from all slip trials (S-1 through S-5, RS-1) and from all trials in the subsequent nonslip block (NS-1 through NS-3) were pooled across subjects for the analysis of backward and forward balance loss, respectively. The goodness of fit for the logistic regression models was assessed by expanding each model with higher order (quadratic and cubic) terms and evaluating the difference in the $-2 \log$ likelihood between the original (reduced) model and the expanded...
The decreased incidence of backward balance loss with repeated slip exposure was related to an increase in stability against backward balance loss at slip onset (i.e., an increase in the value of the predicted stability measure in Fig. 7a). There was significant improvement (P < 0.01) in stability against backward balance loss under slip conditions at seat-off between trials S-1 and S-2 and between trials S-2 and S-3 (Fig. 7a), but no further change for the remainder of the slip block (P > 0.05).

Overcompensation and adaptation

An elevated risk for forward balance loss accompanied the adaptation to repeated slips, due to an equivalent of overcompensation under nonslip conditions. As evidence thereof, the stability at seat-off against forward balance loss under nonslip conditions was significantly less (P < 0.01) in trial NS-1 than in STS (i.e., the value of the predicted stability measure in Fig. 7b was greater), while the stability was not different for NS-1 and the preceding S-5 (P > 0.05).

After only a single nonslip trial (NS-1), subjects adapted to reverse the overcompensation by improving stability against forward balance loss under the nonslip condition (i.e., a significant decrease in the value of the stability measure in Fig. 7b, P < 0.01). This effect was retained with no further changes in stability during the remainder of the nonslip block (from NS-2 to NS-3) and the re-slip trial (RS-1) (P > 0.05). Again, a significant (P < 0.01) logistic relationship existed between predicted stability at seat-off and forward balance loss under nonslip conditions (Fig. 8a). This relationship was sufficient to explain the data, as the addition of higher order terms did not significantly improve the model (P = 0.79). A strong correlation (r = 0.978, P < 0.01) between the estimated and actual incidence of forward balance loss across trials (Fig. 8b) also supports the strength of the model.

Optimal movement strategies

With exposure to slip and nonslip conditions, subjects began to adapt toward an optimal movement strategy that allowed a balance loss to be avoided under both conditions. Such adaptation is demonstrated by the convergence of the COM state at seat-off toward the midline between the loss of balance regions in Fig. 3, a and b. It is important to note that subjects did not return to the regular sit-to-stand behavior that they exhibited prior to the first slip exposure. The adapted subjects were significantly more stable at seat-off against backward balance loss under slip conditions (RS-1 as compared to STS in Fig. 7a) than during the regular STS trial (P < 0.01).

As further demonstration of this optimal movement strategy, the predicted stability at seat-off of the re-slip against a backward balance loss under slip conditions was significantly greater (P < 0.01) than on the first slip (cf. S-1 vs. RS-1 in Fig. 7a). Meanwhile, the predicted stability at seat-off of the re-slip against a forward balance loss under nonslip conditions was greater (P < 0.01) than on the first nonslip trial (cf. NS-1 vs. RS-1 in Fig. 7b). Furthermore, 12 of 38 (31%) older adults analyzed avoided a balance loss in both RS-1 and its preceding NS-3 (an example shown in Fig. 3b) with no differences in predicted stability at seat-off between these trials (P > 0.05 in Fig. 7, a and b). This represents a substantial improvement as compared with 100% backward balance losses in trial S-1 and

![Graph](image-url)
88% forward balance losses in NS-1. The reductions in balance loss were accompanied by a substantial decrease in fall incidence from 73% in S-1 to only 20% in RS-1.

**Discussion**

**Implication, limitations, and predictability**

The experimental results can be fully accounted for if we assume that probability of balance loss and dynamic stability limits under slip and nonslip conditions are predictable mathematically, and perhaps neurophysiologically, and that the feedforward stability control that the CNS employs must require an internal representation of these limits. First, an improvement associated with adaptation to repeated slip exposure in the mathematically predicted stability at seat-off (slip onset) correlated significantly with a reduction in backward balance loss after seat-off during subsequent recovery response to the slip. Second, an overcompensation-related reduction in stability against forward balance loss was associated with an elevated risk for forward balance loss when slips stopped occurring. A subsequent improvement in stability against forward balance loss at seat-off correlated significantly with a reduction in forward balance loss in the nonslip trials. Finally, the predicted optimal movement options began to emerge with alternate exposure to slip and nonslip conditions. Without receiving any explicit or implicit instruction on how to adapt, the older adults began to converge their COM state at seat-off toward the optimal region as predicted mathematically. This adaptive process therefore led to a reduced incidence of balance loss, regardless of whether or not a slip occurred.

The persistence of these proactive, feedforward control adaptations and of any underlying refinements in an internal representation of stability limits is presently unknown, nor is it known whether such refinements in stability limits will transfer to altered feedforward control and a reduced likelihood of balance loss during other tasks. Nevertheless, because of the relatively rare real-life occurrence of slips during a sit-to-stand, the present paradigm provided a unique opportunity to observe older adults' natural process of adaptation with minimal bias from prior experience.

Despite the insights gained, limitations exist in the present analyses of stability. The feasible stability regions were based on a simplified representation of the body by a two-link model and the assumption of an infinite slip distance. Motion at the knees and hips can expand the feasible stability region (Iqbal and Pai 2000), while termination of foot movement after a finite slip aids in preventing a backward balance loss, also altering the feasible stability region.

It was further assumed that subjects stepped strictly from necessity because of balance loss. A previous study found that 17% of forward steps and 41% of backward steps following a postural perturbation may have been unnecessary (Pai et al. 2000). Such inherent error in identifying balance loss might explain the consistent overestimation of balance losses as compared with the number of actual balance losses. This overestimation resulted in a notable deviation of the regression equations in Figs. 6b and 8b from lines of unity. Initiating a step while inside the feasible stability region may reflect a reflexive response, ill-perceived needs, fall-related fear or anxiety, lack of explicit instruction not to step, or simply a choice with no obvious reasons. The feasible stability region is mathematically established by ruling out systematically all the impossible

**Fig. 6.** A: logistic regression model of the relationship between the mathematically predicted stability under slip conditions at seat-off and the probability of backward balance loss, derived from all slip trials (S-1 through S-5, RS-1, n = 241). The broken vertical line indicates the computed threshold for balance loss derived from the logistic regression equation (i.e., the value of x when y = 0.5). B: percentage of backward balance loss in each of the slip trials as estimated from the logistic regression-based balance loss threshold is strongly correlated ($r^2 = 0.957$, $P < 0.01$) with the actual percentage of backward balance loss.
proaches to movement stability include grasping (Holliday et al. 1990; Luchies et al. 1994), ankle and hip motion (ankle/hip strategy) (Horak 1992; Horak and Nashner 1986), and compensatory stepping (Maki and McIroy 1997). The stepping response has a unique and irreplaceable importance in fall prevention, particularly following large disturbances of balance. Arguably, proactive adaptations to movement stability represent a first line of defense against falling, whereas reactive responses represent a second line of defense; both play an important role.

Aging and adaptability in balance control

The present older adults were more likely than young to fall on initial exposure to a slip during a sit-to-stand (Pavol et al. 2002b), yet the mechanisms of falling were similar (Pavol et al. 2002c,d). Subsequently, on repeated slip exposure, older and young adults clearly evidenced similar patterns of adaptive changes in the feedforward control of the COM state trajectory, influencing the likelihood of both balance loss and falls (Pavol and Pai 2002; Pavol et al. 2002d). Adaptive feedforward control of stability based on a continuously updated internal model thus appears to be used by old and young alike. Evidence suggests, however, that the effective size (quantifiable by the triggering threshold of a stepping response) of the feasible stability region decreases with older age, and with it the magnitude of the adaptive changes in feedforward control (Pavol et al. 2002d).

The results indicated that adaptation of the feedforward control began immediately on a change in conditions and that a steady-state adaptation was attained in two trials or less. Such rapid adaptive behavior in feedforward control has also been demonstrated in other activities (Lang and Bastian 1999; Marigold and Patla 2001; Owings et al. 2001; Scheidt et al. 2001). Similarly, a person’s reactive response can be rapidly modulated to better restore balance and upright posture (Buchanan and Horak 1999; Marigold and Patla 2001; Nashner 1976) or to provide better weight support from the slipping limb during the recovery (Pavol et al. 2002d). The fact that fall incidence decreased at a faster rate than the reduction in balance loss is noteworthy. It suggests that slip-related falls decreased, not only because older adults proactively improved their movement stability at slip onset (Fig. 7), but also because they rapidly learned to adaptively improve their reactive response so that the proportion of balance losses resulting in falls decreases.
Such adaptive improvements in reactive response have, in fact, been reported and are similar in older and young adults (Pavol et al. 2002d).

Slips contribute to 25% of falls by older adults (Hausdorff et al. 1997) and often lead to a backward fall incident (Topper et al. 1993) that predisposes the faller to hip fracture (Smeets et al. 2001). It is promising that repeated slip exposure under a protective environment appeared to be effective in facilitating an update or modification of the internal representation of stability limits and in inducing improvements in the feedforward control of movement stability, including the adoption of optimal movement strategies. With such optimal movement strategies, a balance loss can be avoided regardless of whether or not a slip occurs, thereby reducing the reliance on precise or detailed knowledge (which is frequently absent in real-life situations) of forthcoming balance perturbations. It is conceivable that older adults could learn to proactively adopt a similar optimal movement strategy in response to generalized environmental cues, such as a general knowledge of icy weather conditions or a wet floor surface, thereby averting unintended balance losses through feedforward control of stability without sacrificing their mobility.

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DISCLOSURES

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