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Mechanisms Leading to a Fall From an Induced Trip in Healthy Older Adults

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Background. Tripping is a leading cause of falls in older adults, often resulting in serious injury. Although the requirements for recovery from a trip are well characterized, the mechanisms whereby trips by older adults actually result in falls are not known. This study sought to identify such mechanisms.

Methods. Trips were induced during gait in 79 healthy, community-dwelling, safety-harnessed, older adults (50 women) using a concealed, mechanical obstacle. Kinematic and kinetic variables describing the recovery attempts were compared between those who fell and those who recovered. Subjects were analyzed according to the recovery strategy employed (lowering vs elevating) and the time of the "fall" (during step vs after step).

Results. Three apparent mechanisms of falling were identified. For a lowering strategy, during-step falls were associated with a faster walking speed at the time of the trip (91% ± 8% vs 68% ± 11% body height [bh] per second; $p <$ .001) and delayed support limb loading (267 ± 49 milliseconds vs 160 ± 39 milliseconds; $p <$ .001). After-step falls were associated with a more anterior head-arms-torso center of mass at the time of the trip (6.2 ± 1.3 degrees vs 0.2 ± 4.4 degrees; $p <$ .01), followed by excessive lumbar flexion and buckling of the recovery limb. The elevating strategy fall was associated with a faster walking speed (93% vs 68% ± 11% bh per second; $p <$ .001) followed by excessive lumbar flexion.

Conclusions. Walking quickly may be the greatest cause of falling following a trip in healthy older adults. An anterior body mass carriage, accompanied by back and knee extensor weakness, may also lead to falls following a trip. Deficient stepping responses did not contribute to the falls.

TRIPPING is a leading cause of falls in community-dwelling older adults, responsible for up to 53% of falls in this population (1). These falls have serious consequences. Eleven percent of all falls by older adults result in serious injury (2), and falls are the leading cause of unintentional-injury death in older adults in the United States (3). Trip-related falls are specifically responsible for 12% to 22% of the hip fractures suffered by older adults (4,5). Even noninjurious falls can bring about decreased quality of life through the fear of falling and, in turn, the restriction of activities (2,6). There is, therefore, a need for effective interventions for reducing the incidence of trip-related falls in the older adult population.

Studies of fall epidemiology have identified the characteristics of older adults who are most likely to suffer a fall (7), but these studies have not considered the biomechanics of falling. Therefore, factors that are directly related to the ability to prevent a fall have not been differentiated from factors that simply covary with the likelihood of falling. Such a differentiation is needed to appropriately target interventions for fall prevention at the former versus the latter factors.

To better understand the factors directly involved in restoring balance following a trip, studies have characterized the kinematic, kinetic, and neuromotor responses associated with recovering from an induced trip or stumble (8–14). The following three common strategies for recovery have been identified (9). In a lowering strategy, the tripped foot is immediately lowered to the ground on the near side of the obstacle. The tripped limb then acts as the support limb as the contralateral recovery limb executes the initial recovery step across the obstacle. In an elevating strategy or in a reaching strategy, the tripped limb is used as the recovery limb as the tripped foot is lifted over the obstacle in a continuation of the original step. The contralateral stance limb acts as the support limb during the recovery step. Elevating and reaching strategies are differentiated based on whether recovery limb flexion occurs at multiple joints or primarily at the hip, respectively.

Independent of the strategy employed, successful recovery from a trip has been associated, conceptually, with the ability to react rapidly with an appropriate response (15,16) to control the forward rotation of the trunk (12,13) and execute a recovery step of sufficient length to establish a new, functional, base of support (12,17). Also important is the effective use of the support limb in slowing the fall of the head, arms, and torso (HAT) during the stepping phase; that is, from the time of the trip until the recovery foot ground contact (8,9). Finally, recovery requires that the recovery limb provide sufficient hip height during stance for the support limb to execute an effective follow-through step.

Although the requirements for recovery from a trip are fairly well characterized, the primary mechanisms whereby trips by older adults actually result in falls are not known.

To date, the biomechanics of a failed recovery from a trip or stumble have not been reported. Therefore, any fall-prevention efforts aimed at reducing an older adult's likelihood of falling following a trip can, at present, only have been based on theories as to why these falls occur. Since one may hypothesize any number of manners in which the recovery process may fail, many or most of which may rarely be observed in practice, an experimental validation of these theories is clearly needed.

This study attempted to identify the mechanisms whereby selected healthy older adults fell following an induced trip. Ten possible contributing factors were considered. It was hypothesized that, in comparison to those who recovered, fallers (i) were walking faster at the time of the trip, (ii) had a more forward-oriented HAT center of mass at the time of the trip, (iii) fell faster initially, (iv) selected an inappropriate recovery strategy, (v) were slower in initiating the phases of their recovery, (vi) were less effective at slowing their fall through their stepping phase motor response, (vii) took a shorter recovery step, (viii) took a slower recovery step, (ix) experienced buckling of the recovery limb after ground contact, and (x) experienced greater lumbar flexion.

Methods

Subjects

Fifty women and 29 men (age, 72 ± 5 years; height, 1.64 ± 0.09 m; mass, 76.0 ± 14.0 kg), all healthy, community-dwelling, and at least 65 years of age, provided written informed consent to participate in this experiment, which was part of a larger study of falling in these older adults. Each subject was screened by a geriatrician for exclusionary factors that included neurological, musculoskeletal, cardiovascular, pulmonary, and cognitive disorders, as well as a history of repeated falling. Five subjects reported having fallen once in the past year because of an external disturbance. A minimum bone mineral density of the femoral neck, assessed by dual-energy x-ray absorptiometry (Hologic QDR 1000, Waltham, MA), of 0.65 $g \cdot cm^{-2}$ was also required. Subjects were paid for their participation.

Experimental Protocol

Subjects were placed in a safety harness and tripped during gait. This previously described protocol (18) is summarized here.

Subjects wore a full-body safety harness that was attached by a pair of dynamic ropes to a bearing on a ceiling-mounted track. Rope lengths were adjusted such that the wrists and knees could not touch the floor. A calibrated load cell (Omega Engineering, Stamford, CT), in series with the dynamic ropes, measured the force exerted on the ropes by the subject. The safety harness did not introduce any meaningful changes in gait (18).

Trips were induced using a concealed, pneumatically driven, metal obstacle. This obstacle would rise 5.1 cm from the floor in approximately 170 milliseconds when manually triggered by the investigator, inducing a trip by obstructing the toe of the shoe of the swing foot during mid-to-late swing. For the trip, a decoy "tripping rope" was also laid across the gait path, 1.5 m before the mechanical obstacle. The rope provided a visible hazard, the purpose of which was to mislead the subject as to the time, location, and mechanism of the trip.

Subjects were informed that a trip would take place during an upcoming, but unspecified, trial. Instructions were to walk at a self-selected, "normal" speed from a designated starting point to a point approximately 7 m distant, looking straight ahead. If tripped, subjects were to recover and continue walking. On a subsequent trial, the obstacle was triggered and a trip induced. Only one attempt was made to trip each subject.

The kinematics of the trip and subsequent recovery attempt were recorded using a six-camera motion capture system (Motion Analysis, Santa Rosa, CA). The cameras, operating at 60 Hz, recorded the motion of 18 hemispherical passive reflective markers applied over selected anatomical landmarks of the bilateral upper and lower limbs, torso, and head. In addition, ground reaction forces and moments were measured by two forceplates (AMTI, Newton, MA), located immediately preceding the obstacle and in the expected region of recovery foot ground contact, respectively. Forceplate and safety harness load cell data were sampled at 1000 Hz in synchrony with the kinematic data.

Data Analysis

Each trip outcome was classified as either a recovery, fall, rope assist, or miss (18). Falls corresponded to the subject being fully supported by the safety harness. Recoveries and rope assists were differentiated based on the integral, over the 1 second following the triggering of the obstacle, of the filtered, load cell, rope force signal from the safety harness. Trip outcomes with less than 5% body weight · second exerted on the ropes were classified as recoveries. Outcomes with larger integrated forces were considered rope assists. Misses resulted when impact with the obstacle did not occur as intended.

The recovery attempts of subjects who were successfully tripped were classified as a lowering, an elevating, or an "other" strategy. Classifications were based on the earlier descriptions of these strategies, except that elevating and reaching strategies were grouped together since they differ only slightly in the mechanics of their recovery step. Two subjects employing an "other" strategy, in which the tripped foot was lowered onto the obstacle, were excluded from analysis.

Six events were of interest in the analysis. The time of the trip was registered as a high-frequency impact artifact in the data of the forceplate in front of the obstacle. For subjects who employed a lowering strategy, the start and end of the large mediolateral shift in the center of pressure on this same forceplate (computed after recursive, fourth-order, Butterworth low-pass filtering of the data at 50 Hz) identified the times of support limb loading and recovery foot toe-off, respectively. Recovery foot ground contact and the follow-through step toe-off were registered by the underlying forceplates. Where initial recovery foot ground contact occurred beyond the forceplates, this event was identified from the foot-marker paths in the kinematic data. Finally, for those trips that resulted in a fall, the "fall" was defined to occur when 50% of the subject's body weight was supported by the safety harness ropes, as indicated by the filtered load cell rope force signal. All event times were validated against the kinematic data.

The analysis of the recovery attempts was based on a simplified two-link model of the body in the sagittal plane (Figure 1), supplemented by measures of trunk kinematics. The lower limb of primary interest was represented as an elastic link from the ankle to the midpoint of the bilateral hip joint centers. The HAT was represented as a single link from the bilateral hip joint centers to the HAT center of mass. Descriptors of the model included the ankle and hip positions, the hip-to-ankle distance (measured three-dimensionally from the ipsilateral hip joint center), the moment arm of the HAT weight about the ankle, the link orientations with respect to vertical, and the rate of change of each measure.

Descriptors were computed from the kinematic data after recursive, fourth-order, Butterworth low-pass filtering of the reflective marker paths. Marker-specific cutoff frequencies, determined by a residual analysis (19), ranged from 5.5 to 7.5 Hz. Ankle positions were represented by the markers on the lateral malleoli. The locations of the hip joint centers and the positions and spatial orientations of the pelvis, trunk, head, upper arm, and forearm segments were computed from the three-dimensional paths of the reflective markers affixed to the greater trochanters and HAT. The computations employed transformations derived from anthropometric measurements and from kinematic data collected during a static initialization trial. Anthropometric measurements were also used to derive subject-specific estimates of body segment mass and center of mass locations (20). These and the body segment kinematics determined the position of the HAT center of mass. All distances were normalized to body height (bh).

In analyzing trunk kinematics, lumbar flexion was computed as the forward rotation of the trunk segment, with respect to the orientation of the pelvis, about an axis at the level of L_3L_4 and parallel to the pelvis mediolateral axis. The Cardan angle approach of Grood and Suntay (21) was employed. Lumbar flexion was defined to be zero for the relative segment orientation observed during quiet standing

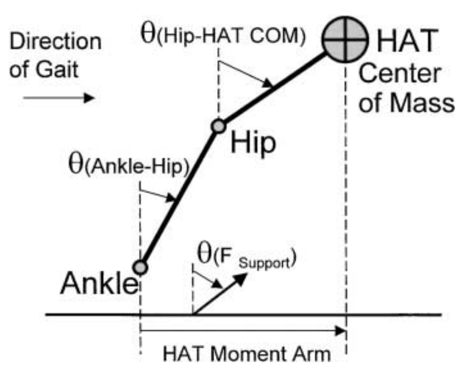

Figure 1. Simplified two-link model of the body in the sagittal plane. The lower limb is modeled as an elastic link from the ankle to the midpoint of the bilateral hip joint centers. The head-arms-torso (HAT) is modeled as a link from the bilateral hip joint centers to the HAT center of mass (COM). All angles (θ) and distances shown are positive, including those for the net support force ($F_{support}$).

with the shoulder joint centers 6.8 cm posterior to the hip joint centers, based on the mode of the observed distribution in our subjects. The forward inclination of the trunk with respect to vertical was computed in the same basic manner, with the vertical orientation of the trunk defined as described previously.

From all possible kinematic variables defined by the previously described events and analytic models, a set of 36 descriptors of the recovery attempts (Tables 1–5) was selected to allow evaluation of the factors hypothesized as contributing to a fall following a trip. One or more variables were uniquely associated with each hypothesized factor (Table 6). Two kinetic variables were also used as gross indicators of the motor responses employed to control the fall during the stepping phase. The support impulse was computed as the integral, from the time of the trip until follow-through step toe-off, of the filtered vertical reaction force measured by the forceplate preceding the obstacle. A propulsive impulse was also computed as the corresponding integral of the filtered anterior-posterior shear force measured by the forceplate, with the direction of the vector formed by the support and propulsive impulses defining the angle of the net support force.

Finally, the phase of gait in which each trip occurred was determined as the perpendicular distance to the obstacle from the static location of the obstructed toe during the preceding stance phase. This distance was expressed as a percentage of the length of the contralateral stride preceding the trip.

Statistics

The kinematic and kinetic variables that differed between those who successfully recovered and those who fell were determined. Subjects were grouped, for this analysis, by the recovery strategy employed. In addition, within this grouping, two distinct groups of fallers who employed a lowering strategy emerged. "During-step" fallers fell within 80 milliseconds of recovery foot ground contact (range, –25 to 77 ms), whereas "after-step" fallers did not fall until after taking a follow-through step (range, 471 to 785 ms after recovery foot ground contact). Because this large difference could reflect differing falling mechanisms, during-step and after-step fallers were analyzed as separate groups. The Mann-Whitney test was used to compare the during-step fallers and the after-step fallers with those who successfully recovered using a lowering strategy. Because only one subject who employed an elevating strategy fell, one-sample *t* tests were employed to determine whether those who recovered using an elevating strategy differed from the elevating faller.

Results of these comparisons were used to determine which of the hypothesized factors contributed to the falls by each of the three groups of fallers analyzed. A factor was considered to have contributed to the falls of a group if any of the kinematic or kinetic variables associated with the factor (Table 6) differed significantly between the fallers and those who successfully recovered using a similar strategy.

Logistic regression analysis was used to determine whether the recovery strategy employed was related to the forward velocity of the hips or the phase of gait at the time of the trip, or

Table 1. Kinematics of the Trip and Initial Fall in Each Group of Subjects

Variable	Lowering Strategy: Recovery (n = 26)	Lowering Strategy: During-Step Fall (n = 5)	Lowering Strategy: After-Step Fall (n = 3)	Elevating Strategy: Recovery (n = 11)	Elevating Strategy: Fall (n = 1)
0. Hip horizontal velocity at time of trip† (m/s)	1.13 ± 0.19	1.45 ± 0.12	1.29 ± 0.09	1.12 ± 0.20	1.40
1. Hip horizontal velocity at time of trip (%bh/s)	68.2 ± 11.1	91.3 ± 7.8****	79.4 ± 6.5	68.0 ± 10.6	92.5****
2. Hip-HAT COM angle at time of trip (degrees)	0.2 ± 4.4	−1.0 ± 2.5	6.2 ± 1.3**	0.6 ± 4.4	1.0
3. Trunk inclination at time of trip (degrees)	9.1 ± 5.7	7.5 ± 3.5	18.8 ± 8.3*	8.7 ± 7.2	14.3
4. Hip horizontal velocity 100 ms posttrip (%bh/s)	72.9 ± 11.0	94.5 ± 5.0****	82.2 ± 13.3	67.4 ± 9.4	86.5****
5. Hip vertical velocity 100 ms posttrip (%bh/s)	−9.8 ± 5.3	−11.8 ± 6.8	−7.2 ± 6.8	−9.3 ± 5.6	−8.1
6. Hip-HAT COM velocity 100 ms posttrip (degrees)	15.9 ± 13.6	22.4 ± 13.9	22.4 ± 13.6	19.5 ± 9.5	42.8****

Notes: Values are mean ± *SD*. Velocities 100 ms posttrip reflect the initial rate of falling and preceded support limb loading in all who employed a lowering strategy. bh = body height; HAT = head-arms-torso; COM = center of mass.

†Variable displayed for informational purposes only; no statistical analyses were performed on or using this variable.

*$p < .05$; **$p < .01$; ****$p < .001$ (vs recovery group for the corresponding strategy).

to the speed or height of the swing ankle 17 milliseconds prior to the trip. The pooled data of the recovery, rope-assist, and fall groups were analyzed. A backward, stepwise, multivariable, logistic regression analysis was subsequently performed. All variables possessing a significant univariate relationship to the recovery strategy were included in the initial model, and the likelihood ratio test with a cutoff probability of 0.1 was used in variable elimination. Outliers (standardized residual greater than 2.0) in the logistic relationship obtained from the stepwise analysis were defined as having employed an inappropriate recovery strategy.

Analyses were performed using SPSS for Windows Release 7.0 (SPSS, Inc., Chicago, IL). A significance level of .05 was used in all analyses except the t tests involving the elevating faller, where a significance level of .001 was used.

Results

Sixty-one subjects were successfully tripped. Forty-three subjects employed a lowering strategy in their recovery attempt, 15 subjects employed an elevating strategy, and 2 subjects (both successful recoveries) lowered the tripped foot onto the obstacle in an "other" strategy. In those who employed a lowering strategy, the outcomes of the trips were 26 recoveries, 5 during-step falls, 3 after-step falls, and 9 rope assists. In those employing an elevating strategy, there were 11 recoveries, 1 fall, and 3 rope assists. Data for one other subject who fell were unavailable, as his trip kinematics failed to record.

The recovery strategy employed was significantly related to the phase of gait in which the trip occurred, with the odds of employing a lowering strategy increasing by a factor of 1.31 (95% confidence interval [CI] 1.08–1.59) for each 1% stride length increase in the phase of gait (Figure 2; $R = .28$, $p = .007$; $n = 58$). In addition, the odds of employing a lowering strategy increased by a factor of 2.05 (95% CI 1.16–3.59) for each 1% bh decrease in the swing ankle height at the time of the trip ($R = –.26$, $p = .013$; $n = 57$). However, the choice of recovery strategy was unrelated to the forward hip velocity ($R = 0$, $p = .241$; $n = 56$) or the swing ankle speed ($R = 0$, $p = .209$; $n = 57$) just prior to the trip. Respective odds ratios were 1.03 (95% CI 0.98–1.08) and 1.01 (95% CI 0.99–1.03) for a 1% bh per second increase in speed. The logistic model obtained through backwards stepwise analysis was identical to the previously described model that included the phase of gait of the trip as the only predictor of recovery strategy. According to our

Table 2. Kinematics and Kinetics of the Support Limb and Head-Arms-Torso During the Stepping Phase in Each Group of Subjects

Variable	Lowering Strategy: Recovery (n = 26)	Lowering Strategy: During-Step Fall (n = 5)	Lowering Strategy: After-Step Fall (n = 3)	Elevating Strategy: Recovery (n = 11)	Elevating Strategy: Fall (n = 1)
7. Time from trip to support limb loading (ms)	160 ± 39	267 ± 49****	144 ± 35	0	0
8. Time from trip to follow-through toe-off (ms)	498 ± 71	490 ± 52	517 ± 75	450 ± 38	400****†
9. Ankle-hip angle at time of loading (degrees)	9.8 ± 4.2	23.6 ± 8.5****	9.1 ± 3.0	8.9 ± 1.8	11.9****
10. Hip-HAT COM angle at time of loading (degrees)	4.4 ± 5.5	16.9 ± 7.8****	10.4 ± 2.6*	0.6 ± 4.4	1.0
11. Moment arm of HAT weight at loading (%bh)	8.7 ± 4.0	23.2 ± 9.5****	9.7 ± 2.5	7.9 ± 1.4	10.2****
12. Lumbar flexion at time of loading (degrees)	6.1 ± 8.9	6.4 ± 4.5	15.4 ± 6.8	6.7 ± 10.9	17.2
13. Support impulse (%bw/s)	44.1 ± 4.5	35.0 ± 6.5***	42.5 ± 3.8	38.3 ± 2.4	31.8****
14. Angle of net support force (degrees)	10.9 ± 2.1	14.1 ± 2.3**	12.1 ± 2.0	10.6 ± 0.8	13.4****
15. Maximum hip upward velocity‡ (%bh/s)	21.4 ± 9.2	3.3 ± 2.7****	5.6 ± 6.5*	14.8 ± 7.2	2.1****

Notes: Values are mean ± *SD*. For those who employed an elevating strategy, the time of support limb loading corresponded to the time of the trip. bh = body height; bw = body weight; HAT = head-arms-torso; COM = center of mass; loading = support limb loading.

†Difference is in direction other than hypothesized.

‡Determined over the period from the time of trip until recovery foot ground contact.

*$p < .05$; **$p < .01$; ***$p < .005$; ****$p < .001$ (vs recovery group for the corresponding strategy).

Table 3. Kinematics of the Recovery Step in Each Group of Subjects

	Lowering Strategy			Elevating Strategy	
Variable	Recovery (n = 26)	During-Step Fall (n = 5)	After-Step Fall (n = 3)	Recovery (n = 11)	Fall (n = 1)
16. Recovery step length‡ (%bh)	49.4 ± 5.7	36.9 ± 8.3****	49.1 ± 8.0	49.8 ± 5.6	51.8
17. Recovery stride length‡ (%bh)	59.9 ± 6.2	51.4 ± 7.2*	61.7 ± 7.4	89.7 ± 5.5	93.2
18. Obstacle-ankle distance at ground contact (%bh)	39.6 ± 5.9	32.0 ± 7.0*	40.0 ± 5.5	32.2 ± 6.8	32.6
19. Minimum hip-ankle distance (%bh)	33.0 ± 3.7	31.8 ± 3.7	28.8 ± 1.4	34.5 ± 2.7	31.0
20. Maximum ankle ground clearance (%bh)	24.7 ± 3.9	23.8 ± 4.9	25.8 ± 2.5	22.1 ± 3.8	24.0
21. Time from trip to recovery foot toe-off (ms)	257 ± 27	280 ± 28	244 ± 10	—	—
22. Time from trip-to-ground contact (ms)	523 ± 44	493 ± 25	505 ± 54	447 ± 46	400
23. Recovery step duration (ms)	265 ± 36	213 ± 43*†	261 ± 63	—	—
24. Maximum horizontal ankle velocity (%bh/s)	263 ± 34	227 ± 50	264 ± 10	225 ± 22	203
25. Average horizontal ankle velocity (%bh/s)	115 ± 15	109 ± 17	117 ± 16	54 ± 9	56
26. Maximum rate of hip-ankle distance decrease (%bh/s)	−144 ± 29	−138 ± 18	−140 ± 23	−80 ± 18	−71
27. Maximum rate of hip-ankle distance increase (%bh/s)	171 ± 30	86 ± 50***	177 ± 8	152 ± 30	134

Notes: Unless otherwise noted, values (mean ± *SD*) were computed for the recovery (i.e., stepping) limb between the times of toe-off and ground contact for a lowering strategy, and between the time of the trip and ground contact for an elevating strategy. bh = body height.

†Difference is in direction other than hypothesized.

‡Computed between the appropriate static positions of the ankles during stance.

*$p < .05$; **$p < .005$; ****$p < .001$ (vs recovery group for the corresponding strategy).

definition, no subject who fell following the trip employed an inappropriate recovery strategy.

As indicated by the kinematic and kinetic descriptors of the recovery attempts, each group of fallers differed significantly in selected aspects from the corresponding group of subjects who recovered (Tables 1–5). These differences were apparent across the entire period from the time of the trip through the time of the fall. Based on the observed differences, 8 of the 10 hypothesized factors were found to play a role in the falls of at least one of the three groups of fallers (Table 6). However, the sets of factors identified as contributing to the falls differed between groups of fallers. For example, a slowed recovery phase initiation contributed only to the during-step falls, as only this group showed a significant difference in one of the three variables associated with this factor in Table 6: the time from the trip to support limb loading (variable 7, shown in Table 2).

Discussion

We have identified three distinct groups of older adults who fell following an induced trip, with observed differences in the factors contributing to these falls suggesting that each group reflected a different mechanism of falling. During-step fallers responded to the trip with a lowering strategy and essentially fell before completing their recovery step. After-step fallers responded using a lowering strategy and were able to successfully execute a recovery step, but proceeded to fall after the subsequent follow-through step. The final faller responded to the trip using an elevating strategy and successfully executed several steps after the recovery step before finally falling.

During-Step Falls

The during-step fallers were walking significantly faster at the time of the trip than those who recovered and, as a result, fell forward faster initially. The during-step fallers also took significantly longer to lower and begin loading the support limb. The combined effect of these factors was that, at the time of support limb loading, the hips and HAT center of mass of the during-step fallers were significantly more forward than in those who recovered (Figure 3A,B). The implications of this difference are seen by considering the direction of the net support force, which is an indicator of the net effect of the joint moments generated over the stepping phase.

Table 4. Kinematics of the Recovery Limb and Head-Arms-Torso at the Time of Recovery Foot Ground Contact in Each Group of Subjects

	Lowering Strategy			Elevating Strategy	
Variable	Recovery (n = 26)	During-Step Fall (n = 5)	After-Step Fall (n = 3)	Recovery (n = 11)	Fall (n = 1)
28. Hip height (%bh)	54.5 ± 2.3	47.2 ± 4.5***	50.9 ± 2.8	54.5 ± 1.5	51.1****
29. Ankle-hip angle (degrees)	−10.1 ± 3.8	12.7 ± 5.9****	−7.8 ± 6.4	−7.6 ± 6.0	0.28
30. Moment arm of HAT weight (%bh)	−2.0 ± 4.8	18.5 ± 4.8****	3.9 ± 4.9*	−0.4 ± 6.5	10.8
31. Hip-HAT COM angle (degrees)	25.6 ± 11.9	41.5 ± 5.2***	38.5 ± 3.1	26.6 ± 8.3	36.7
32. Hip-HAT COM velocity (degrees)	−43.0 ± 52.8	81.6 ± 41.4****	36.1 ± 40.9*	−13.8 ± 57.9	97.3****
33. Trunk inclination from vertical (degrees)	36.0 ± 12.6	48.3 ± 4.7*	55.2 ± 11.4*	37.3 ± 11.1	58.5****
34. Lumbar flexion (degrees)	23.5 ± 10.0	22.1 ± 8.5	38.7 ± 10.9*	23.1 ± 13.3	45.2****

Notes: Values are mean ± *SD*. bh = body height; HAT = head-arms-torso; COM = center of mass.

*$p < .05$; ***$p < .005$; ****$p < .001$ (vs recovery group for the corresponding strategy).

Table 5. Kinematics Related to the Control of the Hips, Recovery Limb, and Trunk Following Recovery Foot Ground Contact for Each Group of Subjects

	Lowering Strategy			Elevating Strategy	
Variable	Recovery ($n = 26$)	During-Step Fall ($n = 5$)	After-Step Fall ($n = 3$)	Recovery ($n = 11$)	Fall ($n = 1$)
35. Maximum hip vertical velocity‡ (%bh/s)	32.2 ± 8.6	—	−0.4 ± 20.0***	29.1 ± 6.4	20.7
36. Minimum hip-ankle distance‡ (%bh)	47.3 ± 2.1	—	41.0 ± 2.2****	47.4 ± 2.2	42.4****
37. Maximum trunk inclination from vertical§ (degrees)	46.6 ± 13.2	47.4 ± 7.0	74.6 ± 26.1*	49.7 ± 12.4	83.5****
38. Maximum lumbar flexion§ (degrees)	35.6 ± 9.1	23.1 ± 10.5*†	54.4 ± 18.7*	35.3 ± 12.2	70.3****

Notes: Values are mean ± *SD*. bh = body height.
†Difference is in direction other than hypothesized.
‡Determined over the 275 ms following recovery foot ground contact, excluding values from after toe-off.
§Determined from the time of trip onward, excluding values from after the time of a fall.
*$p < .05$; ***$p < .005$; ****$p < .001$ (vs recovery group for the corresponding strategy).

In those who recovered, the net support force was directed anterior to the hips and HAT center of mass, potentially allowing it to slow the body's forward rotation during the stepping phase. However, the net support force of the during-step fallers was directed posterior to the hips and HAT center of mass, which would act to accelerate the body's forward rotation. The fallers' delayed support limb loading also resulted in a decreased support impulse. As a probable consequence of these factors, the during-step fallers exhibited a significant deficit in all measures of support limb function. By recovery foot ground contact, the near-continuous forward and downward rotation and translation of the HAT of the during-step fallers had proceeded to the extent that the recovery limb could not be used to prevent a fall (Figure 3E).

Although the recovery step of the during-step fallers was significantly shorter than in those who recovered, a deficient stepping response should not be considered a contributor to these falls. The descriptors in Table 3 suggest that the gross mechanics of the recovery step were similar in the during-step fallers and those who recovered. The shorter step in the fallers, therefore, likely reflects not a deficiency in stepping but a premature recovery foot ground contact caused by the greater pelvic rotation and downward translation observed. In addition, the results (e.g., Figure 3) indicate that only a recovery step much faster and longer than normal might have allowed the during-step fallers to avoid a fall.

After-Step Falls

Similar to the during-step fallers, two of three after-step fallers were walking faster at the time of the trip than all but one subject who recovered. Yet, walking speed was not identified as a significant factor in the after-step falls ($p = .067$), as the third after-step faller was only in the 45th percentile of observed walking speeds. Also in contrast to the during-step fallers, the after-step fallers did not take longer to begin loading the support limb. The after-step fallers did, however, exhibit a HAT center of mass that was oriented significantly more forward of the hips at the time of the trip, due primarily to greater trunk inclination in these individuals. This appears to have played a role in the after-step falls.

Based on the magnitude and direction of the net support force, the after-step fallers and those who recovered employed similar motor responses to control their fall during the stepping phase. However, the after-step fallers continued to exhibit a more forward-oriented HAT center of mass at the time of support limb loading (Figure 3A,C), bringing the HAT center of mass of the after-step fallers closer to their net support force and increasing the gravitational moments of the HAT. These differences would indicate an

Table 6. Results Regarding 10 Factors Hypothesized as Contributing to the Observed Falls

Hypothesized Factor (Manner in Which Fallers Differed)	During-Step Fall	After-Step Fall	Elevating Fall	Associated Variables
Walking faster at time of trip	Yes	No§	Yes	1
More forward-oriented HAT center of mass at time of trip	No	Yes	No	2
Fell faster initially	Yes	No§	Yes	4–6
Inappropriate recovery strategy	No	No	No	†
Slower in initiating recovery phases	Yes	No	No	7, 8, 21
Stepping phase motor response less effective in slowing the fall	Yes	Yes	Yes	13, 15, 28, 31, 32
Took a shorter recovery step	Yes‡	No	No	16–18
Took a slower recovery step	No	No	No	22–25
Recovery limb buckled after ground contact	—	Yes	Yes	35, 36
Experienced greater lumbar flexion	No	Yes	Yes	12, 34, 38

Notes: Each entry indicates whether any of the associated variables listed differed significantly between the designated group of fallers and the recovery group for the corresponding strategy. Variable numbers refer to those in Tables 1–5. HAT = head-arms-torso.
†An inappropriate strategy is evidenced by an outlier in the relationship of Figure 2.
‡Factor may not have contributed to the falls (see "During-Step Falls" in Discussion section).
§Factor may have contributed to some falls (see "After-Step Falls" in Discussion section).

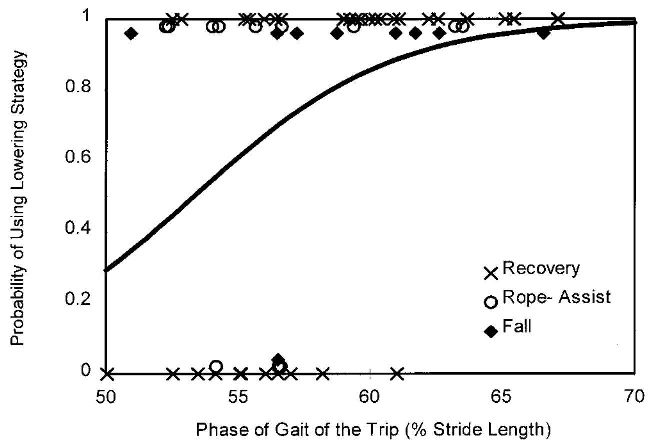

Figure 2. The recovery strategy employed was dependent on the phase of gait in which the trip occurred. Data for the phase of gait of the trip, the recovery strategy employed, and the trip outcome are shown for each subject. Those employing a lowering strategy (n = 43) or an elevating strategy (n = 15) appear across the top and bottom, respectively. The solid line indicates the predicted probability of using the lowering strategy according to the logistic model: $p_{LOWER} = 1/[1 + \exp(-0.2677\theta + 14.277)]$, where θ is the phase of gait of the trip (% stride length). No subject who fell following the trip was classified as an outlier in the depicted relationship.

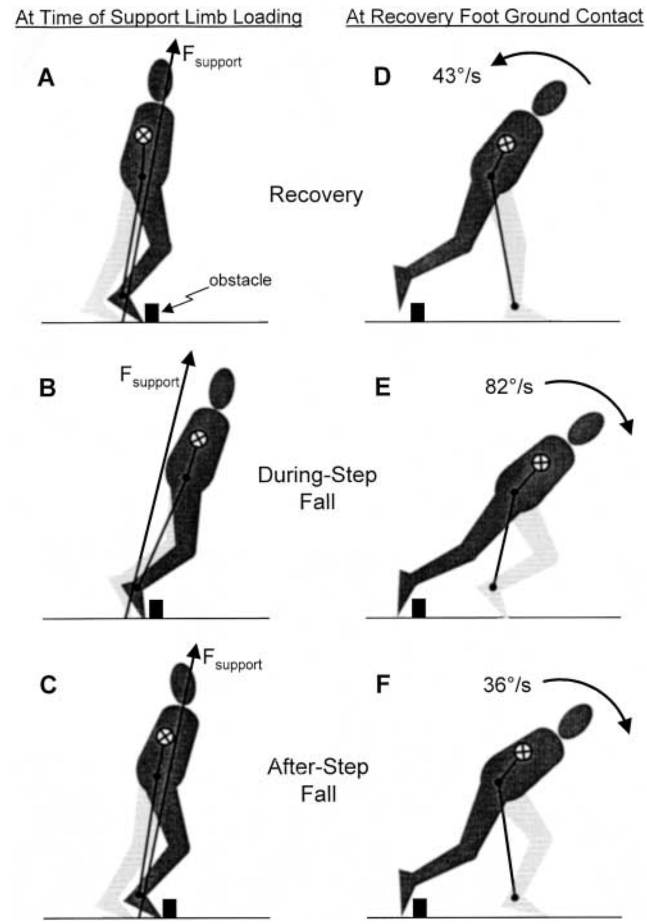

Figure 3. **A–C**, illustrations of the average body states at the times of support limb loading and, **D–F**, recovery foot ground contact for the subjects who successfully recovered using a lowering strategy (shown in **A** and **D**), during-step fallers (shown in **B** and **E**), and after-step fallers (shown in **C** and **F**). The support and recovery limbs are in dark and light grey, respectively. Superimposed is the two-link model used in the analysis. The trunk inclination and the positions of the ankles, hips, and head-arms-torso (HAT) center of mass are as observed. In **A–C**, the net support force ($F_{support}$) is shown emanating in the computed direction from the mean force-weighted average position of the center of pressure. The instantaneous angular velocity of the HAT center of mass about the hips is shown in **D–F**.

impaired ability to slow the HAT's forward rotation. Moreover, this impairment would be magnified in those after-step fallers walking faster at the time of the trip, as the HAT center of mass would pass anterior to the support force sooner and more rapidly.

The probable consequences of these factors were seen at recovery foot ground contact (Figure 3D,F). There were no differences between the after-step fallers and those who recovered in the length, timing, and gross mechanics of the recovery step, nor in the orientation of the hips relative to the recovery ankle at ground contact. Nevertheless, the after-step fallers had experienced greater lumbar flexion and greater forward motion of their HAT center of mass than those who recovered. In addition, the HAT center of mass of the after-step fallers was still rotating forward at recovery foot ground contact, whereas almost all of those who recovered had reversed this rotation. Finally, the after-step fallers were less effective in slowing the descent of their hips during the stepping phase, such that two of these three fallers exhibited lower hip heights at recovery foot ground contact than any subjects who recovered.

The after-step fallers' motor responses to control their fall during the stepping phase resulted in a disadvantaged, more unstable body state at recovery foot ground contact. Nevertheless, three other individuals who used a lowering strategy were in similar states at ground contact and were able to recover, indicating that additional factors led to the observed falls.

To avoid falling after recovery foot ground contact, the after-step fallers needed to arrest their lumbar flexion and the forward rotation of their HAT center of mass before these reached the point of terminal instability. Simultaneously, they needed to prevent buckling of the recovery limb to maintain their hips at a sufficient height to allow for an effective follow-through step. The after-step fallers appear to have failed in both these aspects. They continued in their lumbar flexion and forward trunk rotation to a significantly greater extent than did those who recovered, their recovery limb exhibited significant buckling, and their hips proceeded to drop from an already low height at recovery foot ground contact. The only faller who was able to momentarily reverse the drop of her hips (for 270 milliseconds) raised them well behind her HAT center of mass, thereby rotating her HAT even further forward.

Although the after-step fallers did complete a follow-through step before falling, it was no more effective than their initial recovery step. Their follow-through ankle was 4.4 ± 7.1 degrees ahead of their hips at ground contact, but still 2.0% ± 3.3% bh behind their torso center of mass (see Figure 3F). This would signal a continued inability to restabilize the HAT, the forward orientation of which had not changed, on average, during the step while the hips had dropped by 3.9% ± 2.8% bh. Thereafter, the combined lumbar flexion, forward-oriented HAT center of mass, and downward hip motion inevitably led to the after-step falls.

Elevating Falls

The elevating faller was a hybrid of the during-step and after-step fallers. She was walking significantly faster at the time of the trip than those who recovered, hence was falling forward faster initially. But also, because of the length of her steps and the phase of gait in which the trip occurred, her hips and HAT center of mass were significantly more forward of her support ankle at the time of the trip than in those who recovered.

The consequences of the faster rate of falling and more forward HAT center of mass were similar to those observed in the other groups of fallers. As a result, at recovery foot ground contact, the elevating faller was in a state similar to the after-step fallers. This was despite a more horizontally directed net support force, which should have assisted in slowing the forward rotation of her HAT, and a recovery step that did not differ from that of those who recovered. Beyond recovery foot ground contact, the elevating faller was able to maintain her hips at a sufficient height to allow several steps to be taken. However, her lumbar flexion proceeded to the degree where she was unable to prevent a progressive forward and downward rotation and translation of her HAT to the point of falling.

The mechanisms identified as leading to the falls in each of these three groups of fallers appear to comprise two components, one related to the body's state at the time of the trip and the other related to a deficiency in executing the selected recovery strategy after the trip. It is probable that the observed falls would not have occurred without the presence of both of these components.

In the during-step and elevating falls, the initial component leading to the fall was a faster walking speed at the time of the trip (Table 1). This factor also apparently contributed to two of three after-step falls. Walking quickly may thus be the single greatest cause of falling following a trip in older adults. The contribution of walking speed to the observed falls is consistent with our previous finding that 90% of the trip outcomes in these older adults could be correctly classified using a model that associated longer steps and faster cadences with an increased risk of falling (22). "Hurrying too much" was also the most often-cited reason for falling in a population of community-dwelling older adults (23). The present study provides biomechanical justifications for these past observations. In addition, the results suggest that the most effective means by which older adults can reduce their risk of falling following a trip is by not hurrying when walking.

We note, however, that epidemiological studies invariably associate slower gait with increased falling in older adults, implying that walking speed affects the risk of falling following a trip differently than it affects the risk of tripping or of suffering a non-trip–related fall (22). Slower walking may, therefore, be effective in preventing only trip-related falls, possibly only in those older adults who walk quickly. To this effect, the risk of falling following a trip may be less dependent on absolute walking speed than on speed relative to body height. The walking speeds of the during-step and elevating fallers at the time of the trip did not exceed the adult norm of 1.41 ± 0.16 m per second (24). However, relative to their body heights, these fallers were walking significantly faster ($t = 1.92$, $p < .05$, $df = 50$) than the norm of 83.7% ± 9.8% bh per second for adults aged 20–59 years (24). Furthermore, in our previous analyses, step time and height-normalized step length were selected over absolute walking speed as the best predictors of trip outcome (22).

The common initial component of all after-step falls was not walking speed, but a more forward-oriented HAT center of mass at the time of the trip. This resulted from some combination of a hunched (kyphotic) posture during gait and a more anterior body mass carriage. Since these factors appear capable of facilitating a fall, even during unhurried gait, it may be appropriate to address them in the selected older adults with a hunched posture or anterior mass carriage on the order of our after-step fallers. Note that a similar risk could also be associated with anteriorly carried external loads. Although a hunched posture could also facilitate a fall by reducing the HAT mass moment of inertia, this effect appears to have contributed little to the after-step falls, based on the similar rates of falling 100 milliseconds post-trip in the after-step fallers and those who recovered.

Although two different factors served as the initial component of the mechanisms leading to the observed falls, their effects were similar. All groups of fallers were less able to slow their fall during the stepping phase of their recovery attempts. In the after-step and elevating fallers, the decreased effectiveness of the stepping phase motor response resulted in a disadvantaged body state at recovery foot ground contact. In the during-step fallers, the corresponding result was a fall, due to an additional influence during the stepping phase of the second, post-trip component of their falling mechanism.

The post-trip component leading to the during-step falls was a delay in lowering and loading the support limb. The lowering of the support limb in response to a trip appears to be governed by phase-dependent, polysynaptic spinal reflexes, modulated by supraspinal input (8,9,11). For unknown reasons, this reflex-controlled action was consistently, and almost exclusively, delayed in those individuals with the fastest walking speeds. The delayed time of loading was not a general effect of walking speed, as these variables were only moderately correlated ($r = .43$, $p = .012$). Perhaps the during-step fallers inappropriately chose to employ a voluntary lowering response following an initial elevating reflex. Alternatively, an appropriate lowering reflex may have had a delayed kinematic response due to the post-trip mechanics of the support limb. It does appear that the support ankle attained a greater maximum height following the trip in the during-step fallers. Until the origin of the delay in loading the support limb can be identified, it is unknown whether this factor is amenable to intervention.

The post-trip component of the mechanisms of the after-step and elevating falls was an inability to simultaneously restabilize the trunk and reverse the descent of the hips, in large part because of excessive lumbar flexion and a buckling of the recovery limb. These results with respect to lumbar flexion validate the assertions of Grabiner and colleagues (12) that control of the trunk is a critical factor in recovering from a trip (although the during-step fallers illustrate that limiting lumbar flexion does not guarantee recovery). The diminished control of lumbar flexion in the after-step and elevating fallers could reasonably reflect inadequate back extension strength or power. Similarly, the buckling of the recovery limb in these fallers could reflect inadequate knee extensor strength or power. Such contentions would be consistent with the decreased muscle strength usually observed in community-dwelling (25) and institutionalized (26) older adults with a history of falling. If this is indeed the case, strength training might be an effective intervention in reducing the incidence of trip-related falls in these older adults.

On the other hand, the strength demands of recovery following a trip depend on the motor response employed. An inefficient or inappropriate motor response could cause even strong individuals to fall. Alternately, a stereotyped motor response appropriate for recovery under most circumstances could fail for an increased walking speed or trunk inclination. Such a failure in the after-step and elevating fallers is plausible. The initial motor response to a trip is primarily reflexive (8,9,11,14), thus fairly stereotypical in nature, as is supported by the similar support limb kinetics in the after-step fallers and those who recovered. This and previous studies (12,13) have also found the recovery step to vary little across individuals and walking speeds. Thus, for a given recovery strategy, differences in motor responses may be minor until recovery foot ground contact, by which time the trip outcome is essentially determined. The observed falls could therefore reflect the failure of a "normal" motor response due to the subject's "abnormal" state at the time of the trip. If so, reducing the risk of falling following a trip may be entirely dependent on avoiding these high-risk states.

It is interesting that the only hypothesized factors that appeared not to contribute to the observed falls were those related to a deficient ability to execute the recovery step, given that most studies of fall avoidance in older adults have concentrated on stepping ability (17,27,28). Instead, the ability to effectively use the support limb to slow the fall appears to be of greater importance in determining the outcome of a trip. As observed earlier, poor use of the support limb will result in a severely disadvantaged body state at recovery foot ground contact, from which a fall is either very likely (after-step and elevating fallers) or inevitable (during-step fallers). Yet, the role played by the support limb has been underappreciated to date. Increased study of support limb function is suggested.

There are notable limitations to this study. The small number of subjects in each group of fallers allowed only large differences in the kinematic and kinetic variables to be detected. There may therefore be additional, lesser differences contributing to the observed falls. In addition, we assumed in the analysis that a common mechanism governed the falls in each group. Such an approach was considered valid in establishing general mechanisms of falling from a trip. There were, however, subtle, as-yet-unexplored differences between the falls within a group. Because of the exploratory nature of this study, no single factor could be investigated in detail.

Finally, the mechanisms of falling identified are those that were most commonly observed in a population of healthy community-dwelling older adults, most less than 80 years of age and none with a history of repeated falling. There are likely other mechanisms of falling from a trip across the general population of older adults and the relative incidence of falls due to these various mechanisms may vary across different subpopulations. Nevertheless, it is reasonable that the mechanisms that have been identified could potentially apply to all older adults.

In conclusion, three apparent mechanisms whereby a trip by an older adult can lead to a fall have been identified. During-step falls were associated with a faster walking speed at the time of the trip and a delay in support limb loading. After-step falls were associated with a more anterior HAT center of mass at the time of the trip, followed by excessive lumbar flexion and a buckling of the recovery limb during the recovery attempt. The elevating fall was associated with a faster walking speed followed by excessive lumbar flexion. Overall, the results indicate that walking quickly may be the greatest cause of falling following a trip in healthy older adults.

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