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Impact severity in self-initiated sits and falls associates with center-of-gravity excursion during descent

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Abstract

Although the energy available during a fall from standing greatly exceeds that required to produce hip fracture, this occurs in only about 2% of falls in the elderly. This is thought to be due in part to one's ability to reduce the vertical impact velocity (v_v) and kinetic energy (KE_v) of the body through energy absorption in the lower extremity muscles during descent. The present study tested the hypothesis that the magnitude and percent attenuation in v_v and KE_v associate with the horizontal and vertical excursion of the body's center-of-gravity during descent. Measures were acquired of whole-body kinematics and lower extremity kinetics as young subjects underwent backward descents involving vertical drops of either thigh length (SIT) or lower extremity length (FALL), and horizontal pelvis excursions of either 33 or 66% of lower extremity length. In all trials, subjects attempted to "land as softly as possible." While attenuation in v_v and KE_v (which averaged 62 and 92% respectively), did not associate with trial type, raw magnitudes of these parameters did, with v_v averaging 2-fold greater, and KE_v averaging 6-fold greater, in 66% FALL than in 33% SIT or 66% SIT trials. This was due to a rapid increase in downward velocity accompanying the final stage of descent in 66% SIT and 66% FALL trials, which coincided with the knee moving posterior to the ankle. Accordingly, severe impacts likely accompany not only large fall heights, but also falls where the feet are thrown rapidly forward, as during a backward slip. © 2000 Elsevier Science Ltd. All rights reserved.

Keywords: Falls; Balance; Hip fracture; Joint kinetics; Impact

1. Introduction

Hip fractures represent an enormous health problem in the elderly, and approximately 90% of these are caused by falls (Spaite et al., 1990; Grisso et al., 1991). Yet only 1–2% of all falls in the elderly result in hip fracture (Tinetti et al., 1988; Sattin, 1992). This is surprising from a biomechanical perspective, given that the energy available during a typical standing-height fall exceeds that required to fracture both the elderly and young proximal femur (Lotz and Hayes, 1990; Robinovitch et al., 1991; Courtney et al., 1995).

Several mechanisms have been identified to account for this discrepancy (Cummings and Nevitt, 1989). For example, evidence exists that during an unexpected fall, young subjects avoid direct impact to the hip region, and distribute the body's impact energy by impacting the outstretched hand and pelvis near-simultaneously (Hsiao and Robinovitch, 1998). Furthermore, data suggest that in the event of impact to the hip, substantial energy is absorbed by the skin and fat overlying the hip region, thereby lowering the energy that the underlying bone must absorb (Lauritzen and Askegaard, 1992; Robinovitch et al., 1996). Finally, experiments with young subjects indicate that during a fall from standing, the body's impact velocity and kinetic energy are below values predicted by simple free-fall assumptions (van den Kroonenberg et al., 1996; Hsiao and Robinovitch, 1998).

The last phenomenon arises presumably from the production during descent of "energy-absorbing" work in the eccentrically contracting lower extremity muscles. The act of sitting illustrates the potential effectiveness of this mechanism. During a stand-to-sit movement, one's potential energy decreases typically by about 200 J, but one impacts the chair with relatively low velocity and little risk for injury.

Improved understanding of the biomechanical and neuromuscular variables which govern this mechanism

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would enchance our ability to (a) identify the factors which separate injurious and non injurious falls, and (b) design exercise-based therapies to enhance safe-landing abilities. From a biomechanical perspective, it seems likely that impact severity (as defined by the vertical velocity and kinetic energy of the body at impact) will increase with increases in the vertical descent distance (and potential energy) of the fall. It also seems likely that impact severity will increase with increases in one's degree of imbalance (i.e., the horizontal excursion of the center-of-gravity (COG) with respect to the ankles) during descent. To test whether these supposed relations exist, we examined whole-body kinematics and lower extremity kinetics during self-initiated (volitional) descents from standing to pelvis impact targets that resided at either knee height ("SIT" trials) or ground level ("FALL" trials), and at horizontal distances from the ankles equal to either 33 or 66% of lower extremity length. The specific hypotheses we tested was that the magnitude and percent attenuation in the vertical velocity of the pelvis and kinetic energy of the entire body at the moment of impact associate with the horizontal and vertical excursion of the body's COG during descent.

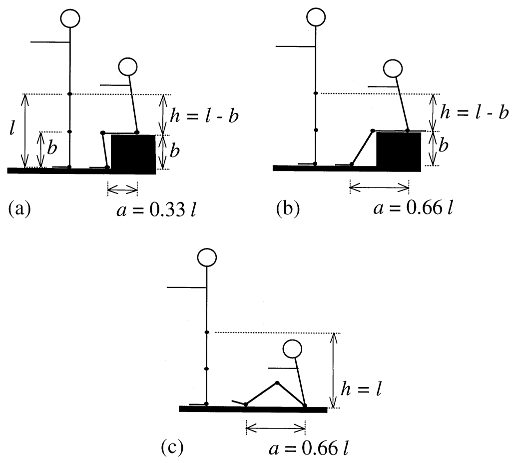

Fig. 1. Schematic of (a) 33% SIT trials; (b) 66% SIT trials; and (c) 66% FALL trials. In all trials, subjects were instructed to "move from a standing to sitting position, and to impact the mat softly."

2. Material and methods

Five females and four males participated, of mean age 27 ± 4 (S.D.) yr (range: 19–36 yr). mean body mass 64 ± 17 kg (range: 49–98 kg, and mean height 1.66 ± 0.12 m (range: 1.52–1.83 m). Each provided informed consent, and the experiment was approved by the Committee on Human Research at the University of California, San Francisco.

Each subject participated in three series of trails, which we shall refer to as 33% SIT, 66% SIT and 66% FALL (Fig. 1). These were distinguished by the horizontal distance between the heels and a 2 cm square impact target painted on a gymnasium mat (set to either 33 or 66% of lower extremity length, defined as the height of the greater trochanter above the floor while standing), and vertical descent distance of the hip (set to either thigh length (SIT) or lower extremity length (FALL)). Each subject first performed 33% SIT trials, then 66% SIT trials, and finally 66% FALL trials. Five repeated trials were acquired in each series. Practice trials were not provided. Rest breaks of 15 s duration were provided between individual trials, and no subject complained of fatigue. Subjects performed the trials barefoot and wore short pants and t-shirts.

Prior to each trial, the subject was instructed to "move from a standing to sitting position, and impact the mat softly." They were also instructed to maintain their arms extended in front of them during descent, and to try to land with the buttocks centered on the impact target. Before commencing their descent, subjects positioned their feet shoulder-width apart, and turned their head to visually inspect the impact target for a 3 s period. However, they maintained their gaze forward throughout descent. No further restrictions were placed on subjects' movement style, or strategy for reducing their impact velocity. Scoring of a successful "hit" was not done, although we noted during the experiments and through data analysis that all trials involved impacts near the target. Furthermore, no trial resulted in lift-off or slipping of the feet before the instant of pelvis impact. Bouncing movements after the initial impact, which were minimal, were not analyzed.

During each trial, a 6-camera motion analysis system (Qualysis Inc., Glastonbury, CT) having a measurement accuracy of approximately 1.0 mm acquired the 3-D positions of 20 markers placed at the head, shoulder, elbows, wrists, anterior-superior-iliac-spines (ASIS), sacrum, knees, ankles, midpoints of the thighs and shins, and third metatarsals. Reaction forces on the subjects's right foot were simultaneously acquired with a force platform (model 4060H, Bertec Corp., Worthington, OH). Muscle activity at the quadriceps, hamstring, tibialis anterior, and soleus was monitored with surface electromyographic (EMG) electrodes, although these data are not reported here. Force and EMG data were sampled at 960 Hz, and motion data were sampled at 60 Hz.

2.1. Data analysis

Stick figure animations from each trial (MacReflex software, Qualysis, Glastonbury, CN) were inspected to detect sit initiation (as the onset of steady vertical or horizontal movement of the hip marker), and confirm that (a) the motion measurement system acquired all body markers during descent; (b) body movements were primarily in the sagittal plane, (c) right and left lower extremity joint rotations were predominantly symmetric, and (d) the body's sagittal plane was nearly aligned with the x–z plane of the motion measurement system. For two subjects, ASIS markers were lost from camera view during 66% FALL trials, reducing the effective number of subjects in that series to seven. All other trials met the above criteria.

Custom routines (MATLAB, The MathWorks, Natick, MA) were used to filter marker position data (fourth-order Butterworth, cut-off frequency = 6 Hz; justified through spectral analysis) and calculate sagittal-plane rotations of individual body segments. Segment velocities and accelerations were then calculated using a forward-difference technique. These data were used to compute the COG location and kinetic energy of the entire body, based on anthropometric relations provided by Dempster (Winter, 1987). They were also input with synchronized records of foot contact force and center-of-pressure (COP) data into an inverse-dynamics routine, which calculated sagittal-plane torques at the hip, knee, and ankle. Joint angles were defined positive if dorsiflexion at the ankle, and flexion at the knee and hip. Joint torques were defined positive if plantarflexor at the ankle, and extensor at the knee and hip. Joint work was defined positive if it was energy-absorbing, i.e., associated with eccentric joint torque.

For each trial, impact severity was expressed by the vertical component of hip velocity (v_v) and the vertical kinetic energy (KE_v) of the entire body at the instant of pelvis impact (see Appendix A for a description of how KE_v was defined). The instant of pelvis impact was estimated as the first time frame where the vertical distance between impact surface and right ASIS marker was less than that observed during static measures of the subjects sitting on a rigid platform. For each trial, we also calculated normalized values of vertical velocity and kinetic energy (which we shall refer to with the symbols v'_v and KE'_v) by dividing raw magnitudes of v_v and KE_v with theoretical estimates of these parameters for a fall involving zero energy absorption in the muscles and ligaments during descent, but the same vertical and horizontal excursions of the COG (see Eqs. (B.6) and (B.7) of Appendix B).

2.2. Statistical analysis

We used repeated-measures analysis of variance (ANOVA) to assess whether v_v, KE_v, v'_v, and KE'_v associated with trial type. Where significant effects were detected, follow-up comparisons were performed through paired t-tests and a Bonferonni correction of $\alpha = 0.01$ to the assumed significance level. All tests were performed with statistical analysis software (SPSS Inc., Chicago, IL).

3. Results

While both v_v and KE_v associated with trial type ($p < 0.001$), each was affected more by the vertical

Table 1
Average parameter values[g]

Parameter	33% SIT ($n = 9$)	66% SIT ($n = 9$)	66% FALL ($n = 7$)
Descent height (m)	0.31 ± 0.05	0.30 ± 0.04	0.72 ± 0.07
Loss in potential energy (J)	193.2 ± 87.6	218.1 ± 104.5	412.3 ± 139.5
COP–COG (m)[ab]	0.07 ± 0.03	0.18 ± 0.05	0.29 ± 0.05
Vertical pelvis velocity (m/s)[bc]	0.45 ± 0.12	0.59 ± 0.16	1.45 ± 0.50
Vertical kinetic energy (J)[b]	4.68 ± 3.58	4.74 ± 3.03	31.56 ± 23.94
Normalized impact velocity[bd]	0.38 ± 0.09	0.37 ± 0.11	0.39 ± 0.13
Normalized vertical kinetic energy[bd]	0.10 ± 0.06	0.06 ± 0.03	0.07 ± 0.03
Ankle plantarflexion (°)[e]	− 3.1 ± 4.9	13.6 ± 4.6	32.9 ± 7.8
Ankle dorsiflexion (°)[e]	5.9 ± 4.2	2.1 ± 2.7	9.1 ± 4.7
Knee flexion (°)[e]	74.8 ± 5.6	51.0 ± 11.7	104.0 ± 14.6
Hip flexion (°)[e]	112.9 ± 12.9	123.6 ± 14.4	175.6 ± 9.5
Ankle plantarflexor torque (N m)[e]	18.1 ± 11.6	17.1 ± 13.2	15.6 ± 11.4
Ankle dorsiflexor torque (N m)[e]	4.7 ± 5.3	7.3 ± 4.9	9.1 ± 3.9
Knee extensor torque (N m)[e]	45.0 ± 16.7	30.1 ± 17.6	46.7 ± 23.3
Hip extensor torque (N m)[e]	53.1 ± 25.0	66.6 ± 25.3	44.1 ± 26.7
Ankle work (J)[bf]	0.4 ± 0.8	1.0 ± 1.7	4.1 ± 2.9
Knee work (J)[bf]	28.9 ± 12.3	10.3 ± 10.8	37.6 ± 26.3
Hip work (J)[bf]	46.0 ± 31.8	59.4 ± 39.4	73.5 ± 27.4
Total joint work (J)[bf]	75.4 ± 36.1	70.8 ± 38.1	115.1 ± 46.1

[a]Horizontal distance between the foot center of pressure and the body's COG.
[b]Magnitude of parameter at instant of pelvis impact.
[c]Downward velocity is positive.
[d]Computed from Eqs. (B.6) and (B.7).
[e]Peak magnitude of parameter during descent.
[f]Absorbing work is positive, generating work is negative.
[g]Cell entries show mean values ± one (S.D.)

than horizontal excursion of the COG during descent (Table 1). Between 33% SIT and 66% SIT trials, the average value of v_v increased by 0.14 m/s (95% C.I. = − 0.02 to 0.29 m/s; $t = 2.0$, df = 8; $p = 0.08$) and the average value of KE_v increased by 0.07 J (95%C.I. = − 2.81 to 2.95 J; $t = -0.1$, df = 8; $p = 0.96$); neither of these increases were statistically significant. In contrast, between 66% FALL and 66% SIT trials, the average value of v_v increased by 0.83 m/s (95% C.I. = 0.39 to 1.28 m/s; $t = 5.6$, df = 6; $p = 0.004$) and the average value of KE_v increased by 26.2 J (95% C.I. = 5.81 to 46.5 J; $t = 3.1$, df = 6; $p = 0.02$).

However, trial type did not associate with normalized indices of impact velocity or kinetic energy ($p = 0.98$ for v_v' and $p = 0.26$ for KE_v'), indicating that the percent attenuation in impact severity was unaffected by both the vertical and horizontal excursion of the COG during descent (Table 1). The average value of v_v' was 0.38, indicating an attenuation of 62% in impact velocity. The average value of KE_v' was 0.08, indicating an attenuation of 92% in kinetic energy at impact.

Strong similarity existed between the three series in movement profiles during the initial stages of descent, which involved first an increase in downward pelvis

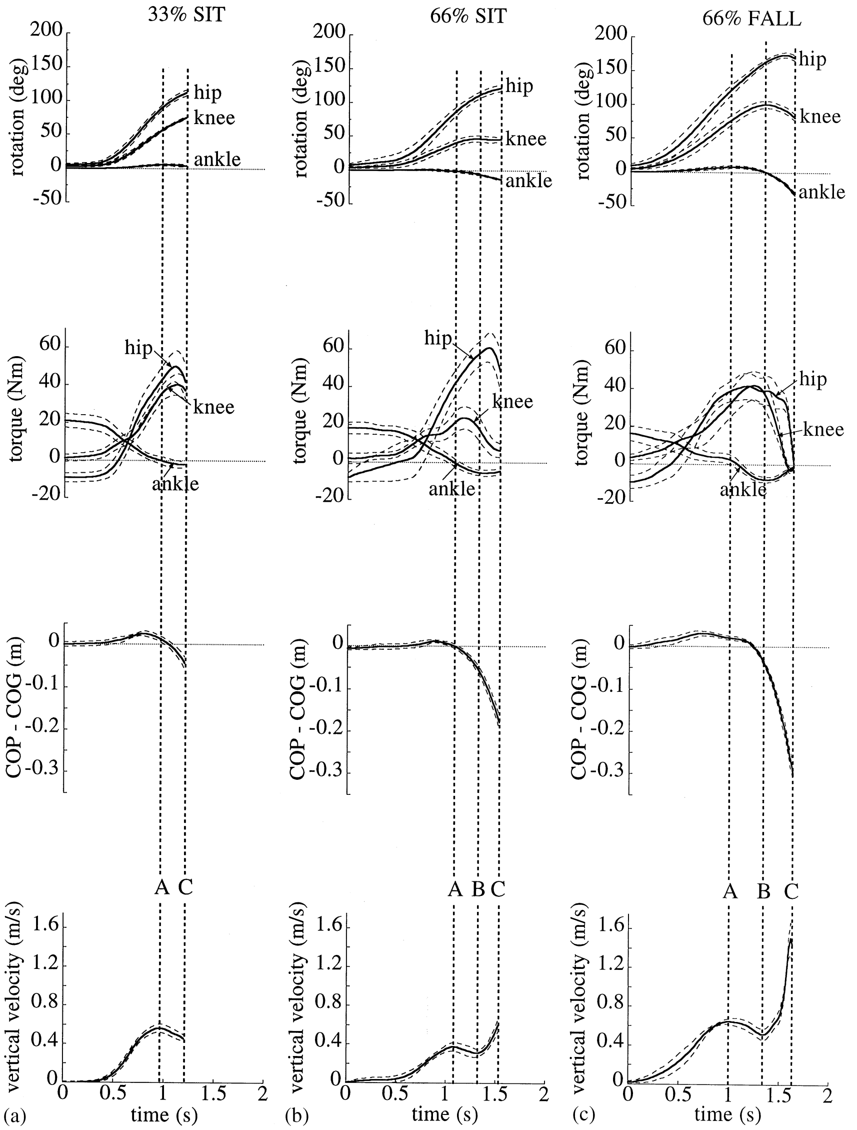

Fig. 2. Temporal variations in joint rotations and torques, horizontal distance between the center-of-gravity and foot center-of-pressure (COG–COP), and vertical velocity of the pelvis during (a) 33% SIT trials, (b) 66% SIT trials, and (c) 66% FALL trials. Solid lines are composite traces based on averaging temporal variations from all subjects, after synchronizing each trial to the instant of pelvis impact. Dashed lines show standard errors. 33% SIT trials were characterized by a single increase and decrease in downward velocity. In contrast, the final portion of descent in 66% SIT and 66% FALL trials was marked by a second increase in downward velocity, which coincided with increased horizontal distance between the COG and COP, movement of the knee behind the ankle, and (especially in the 66% FALL case) a reduction in extensor torques at the knee and hip. A = onset of upward acceleration; B = onset of late-stage downward acceleration; C = pelvis impact. Joint angles are defined positive if dorsiflexion at the ankle, and flexion at the knee and hip. Joint torques are defined positive if plantarflexor at the ankle, and extensor at the knee and hip.

velocity, followed by a period of decreasing downward velocity or upward acceleration (shown as the interval between the lines "A" and "C" in Fig. 2(a), and between the lines "A" and "B" in Figs. 2(b) and (c)). This reflected subjects' ability to generate vertical foot reaction forces during this period which substantially exceeded body weight (Fig. 3). Joint rotations tended to involve gradually increasing flexion of the knees and hips, and increasing and then decreasing dorsiflexion of the ankles (with the knees maintained anterior to the ankles). Corresponding joint torques tended to be eccentric (extensor at the knees and hips, and initially plantarflexor and then dorsiflexor at the ankles), with the exception of a small concentric hip flexor torque at the initiation of descent.

However, while 33% SIT trials tended to terminate during this period of decreasing downward velocity, 66% SIT and 66% FALL trials tended to involve a rapid increase in downward velocity (and decrease in vertical foot reaction forces) during the final stage of descent (Figs. 2 and 3). While this period of "uncontrolled" descent was longer for 66% FALL trials than 66% SIT trials (thus accounting for the greater magnitudes of v_v and KE_v which accompanied the former), its onset in both series appeared to coincide with (a) an increase in the horizontal distance between the COG and COP, (b) a rotation of the ankles into plantarflexion (i.e., movement of the knees behind the ankles), and (c) a reduction in the magnitude of knee and hip extensor torques.

In all series, subjects employed large hip flexions to maintain the trunk inclined forward to the vertical throughout descent (Fig. 4). When compared to 33% SIT trials, 66% SIT and 66% FALL trials were accompanied by increased forward trunk inclination during the middle and final portions of descent, which acted to reduce the horizontal distance between the COG and COP. In all series, hip extensor torques caused a slight rotation of the trunk back towards the vertical just before impact.

Energy-absorbing work at the hip during descent exceeded that at the knee (and ankle) for all three trial types (Table 1 and Fig. 5). Average differences between hip and knee work were 17 J for 33% SIT trials, 49 J for 66% SIT trial, and 35 J for 66% FALL trials. This reflected the occurrence of greater peak flexion rotations and (except in 66% FALL trials) extensor torques at the hip than at the knee. Interestingly, hip and knee extensor torques declined rapidly during the final portion of descent in 66% FALL trials, where these torques became concentric and thus energy generating.

4. Discussion

Our results show that, during both sitting and backwards falling, subjects can substantially reduce their impact velocity and kinetic energy through energy absorption in their lower extremity muscles during descent. In particular our normalized impact severity parameters (v_v' and KE_v') indicate that, in each series, subjects attenuated their vertical impact velocity by approximately 60% and their vertical kinetic energy by 90%. This was achieved primarily through the generation of eccentric extensor torques at the hips and knees, which increased the vertical component of foot reaction force, and thereby decreased downward acceleration.

Our data also indicate that impact severity associates with both the horizontal and vertical excursion of the

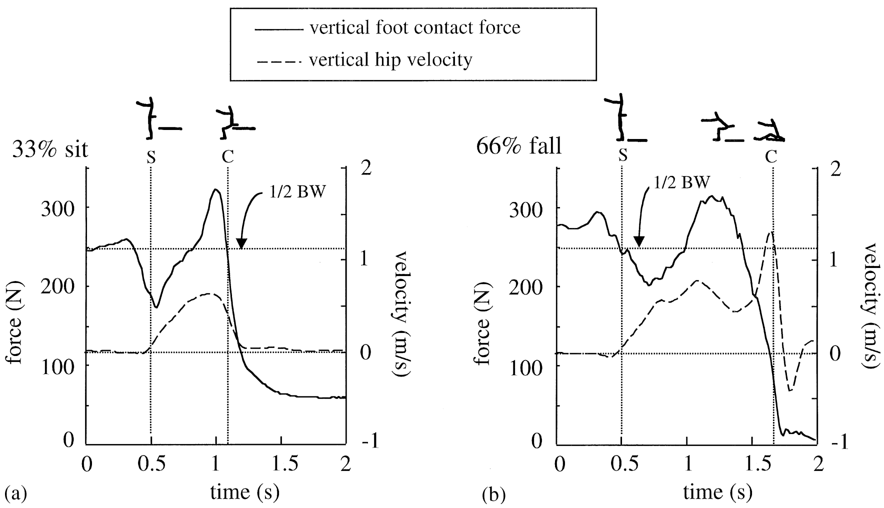

Fig. 3. Association between vertical velocity (dashed lines) and vertical foot reaction forces (solid lines) during individual trials with a male subject. (a) During the final descent stage of 33% SIT trials, foot contact forces tended to exceed body weight, causing a decline in downward velocity. (b) During the final portion of descent in 66% FALL trials, foot contact forces tended to be below body weight, causing downward velocity to increase. Vertical dotted lines show initiation of sit movement (S) and instant of pelvis impact (C); horizontal line shows one-half body weight (250 N). Stick figures at top show this subject's actual body configurations during the trials (anterior pelvis markers are at right and left ASIS, posterior pelvis marker is at L5/S1).

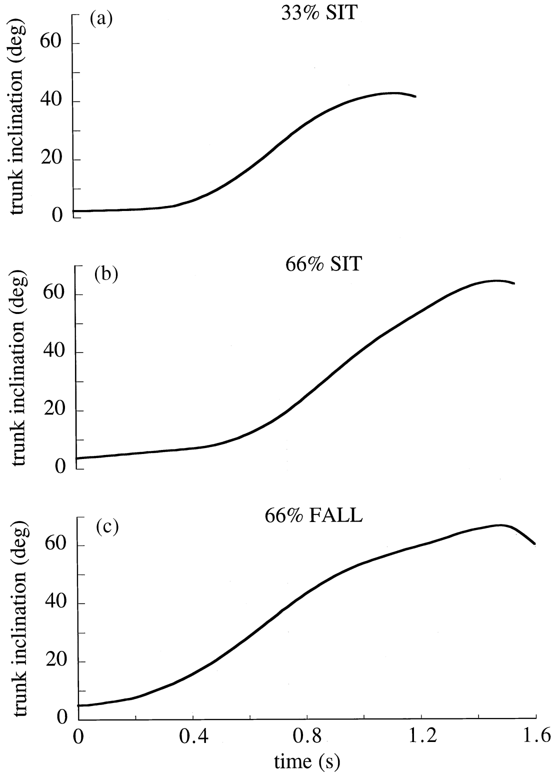

Fig. 4. Average variations in the angular position of the trunk with respect to the vertical during descent for (a) 33% SIT trials, (b) 66% SIT trials, and (c) 66% FALL trials. Data represent composite trends, based on joint rotation shown in Fig. 2. In all series, the trunk was maintained forward to the vertical throughout descent. When compared to 33% SIT trials, 66% SIT and 66% FALL trials were accompanied by increased forward trunk inclination during the middle portion of descent, perhaps reflecting a desire to minimize the horizontal distance (i.e., moment arm) between and COG and COP.

COG during descent. When the former was increased from 33 to 66% leg length, subjects lost their ability to attenuate downward velocity during the final portion of descent. While future experimental and theoretical studies are required to determine the exact cause of this transition, our current data suggest that it relates to both an increase in the horizontal distance between the COG and COP (state of imbalance) and movement of the knee into a position posterior to the ankle. To illustrate the importance of the latter, consider that the vertical velocity of the hip (in fixed space) depends on the angular velocity of both the thigh and shin. When the knee is anterior to the ankle, a knee extensor torque (while likely eccentric, since foot reaction force will tend to flex the knee) reduces downward hip velocity by reducing the downward angular velocity of both the thigh and shin. In this configuration (which persisted throughout descent in the 33% SIT trials), velocity attenuation was possible despite considerable horizontal movement of the COG behind the COP (state of imbalance). In contrast, when the knee is posterior to the ankle an instability arises, where neither a knee extensor nor flexor torque will reduce downward velocity effectively: a flexor torque (while likely eccentric, since foot reaction forces will tend to extend the knee) will reduce the downward angular velocity of the shin, but increase the downward velocity of the thigh. Conversely an extensor torque will decrease the downward velocity of the thigh, while increasing the downward velocity of the shin. In our 66% SIT and 66% FALL experiments, subjects appeared to compensate for this dilemma by simply reducing the magnitude of their knee torque during the final stage of descent.

We found that energy-absorbing work at the hip exceeded that at the knee in all three trial types. This was a direct result of the large hip flexion rotations and (eccentric) hip extensor torques which accompanied descent. Such torques were coordinated (along with those at the ankles and knees) to maintain the trunk in a forwardly flexed position throughout descent. Based on considerations similar to those just discussed for the knee, this allowed a hip extensor torque to retard downward movement of both the trunk and thigh, and thereby maximize impact severity. Other benefits of maintaining the trunk inclined forward to the vertical include avoiding head impact, and minimizing the horizontal excursion of the COG with respect to the ankles (and thus the moment due to gravitational forces).

Work at the ankle tended to be smaller than work at the knee and hip. However, in all series, subjects generated ankle dorsiflexor torques during the final stage of descent, which had at least two important roles: (1) it moved the COP posteriorly, thus reducing the horizontal distance between the COG and COP; and (2) it offset the tendency of knee extensor torques to rotate the shin posteriorly, and place the knee into an unstable position behind the ankle.

Important limitations exist to this study. First, our participants were young individuals having no significant impairment in lower extremity strength or flexibility. On the one hand, aside from the large hip flexion rotations observed in 66% FALL trials, the peak joint torques and rotations observed in our experiments are generally within the range of previously reported values for healthy elderly (Larsson et al., 1979; Murray et al., 1985; Borges, 1989; Stalberg et al., 1989; Schultz, 1992). However, casual observation suggests that elderly vary greatly in their ability to descend from standing to sitting in a controlled manner, and how this relates to factors such as lower extremity strength and flexibility, or central deficits in motor control, can only be assessed through additional studies. Second, we examined only backwards sits and falls, and again further studies are required to assess whether similar mechanisms govern impact severity during sideways and forward falls. Sideways falls are of particular interest due to the high associated risk for hip fracture (Nevitt and Cummings, 1993; Greenspan et al., 1994), and their mechanics may be particularly complex due to asymmetry between the feet and out-of-plane

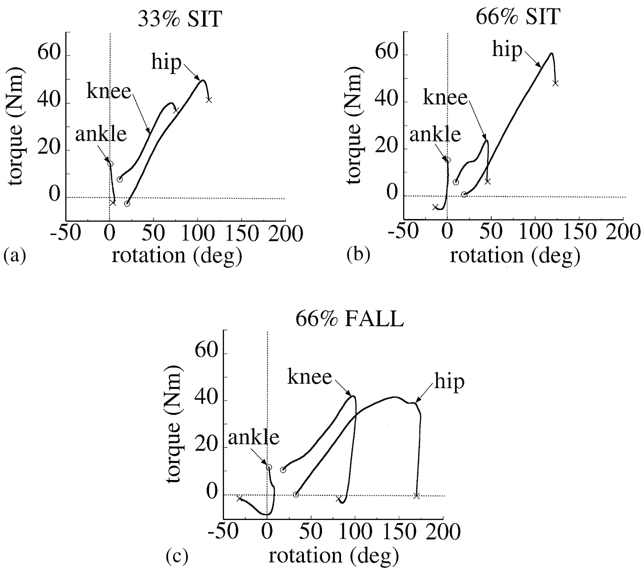

Fig. 5. Average joint torque-rotation behaviour for (a) 33% SIT trials, (b) 66% SIT trials, and (c) 66% FALL trials. Again, data represent composite trends, based on joint rotation and torque data shown in Fig. 2. Open circles indicate commencement of descent, while X's indicate the instant of pelvis impact. For all three trial types, joint work at the hip exceeded that at the knee (and ankle), due to the greater peak extensor torque and flexion rotation generated at this joint (except in 66% FALL trials, where similar peak torques occurred at the hip and knee). Note also that knee and hip extensor torques declined as the joints extended during the final descent stage of 66% SIT and 66% FALL trials, thereby preventing a reduction in net joint work.

rotations (Hsiao and Robinovitch, 1998), Third, our current data might be regarded as describing "best-case" falls, since subjects self-initiated their descent and deliberately attempted to land softly on target impact locations. Nevertheless, we believe that they establish a necessary baseline for examining how impact severity is influenced by factors such as the nature of the perturbation, and "reaction time" delays in the initiation of protective responses. In a previous study, we found that unexpected loss-of-balance elicits attempts to prevent the fall by stepping, and (failing that) to arrange the body into a safe landing configuration (Hsiao and Robinovitch, 1998). However, backwards falls in that study involved qualitatively similar joint rotations to those observed here, and while pelvis impact velocities were higher (at 2.55 ± 0.85 (S.D.) m/s), they were again well-below values predicted by free-fall assumptions. Fourth, it is possible that our results were affected by subject learning during the experiment. However, the small within-subject variability in joint torques and rotations between the five repeated trials in a given series suggests that this was small. It also possible that an order effect occurred due to lack of randomization in the trial sequence. However, we again doubt that this was significant, since it would seem to imply that, among the large number of stand-to-sit movements performed daily, the most recent of these would have an especially profound effect an observed behaviour.

Our results also have important implications regarding the cause and prevention of fall-related injuries. First, they indicate that severe impacts are likely to occur not only during falls from high heights, but also during falls which involve large horizontal excursions of the COG with respect to the ankles. Perturbation conditions which cause the knee to move quickly posterior to the ankle, as one might expect during a backward slip, would appear to create particularly high risk for injury. This supports the importance of non-slip footwear, and the avoidance of slipping hazards such as throw rugs and wet or icy surfaces. Second, our results show that velocity attenuation during a fall, rather than depending on the flexibility or strength of a single joint, results from a synergistic pattern of torques and rotations at the ankle, knee, and hip. Among the motor functions important to this pattern are the flexibility of the ankle in dorsiflexion and the knee and hip in flexion, and the strength of ankle dorsiflexor and knee and hip extensor muscles. Accordingly, each of these represent reasonable targets for exercise therapies aimed at preventing fall-related injuries. Finally, our data indicate that sitting movements (and especially those involving large horizontal excursions of the COG) represent reasonable models of backwards falling. Accordingly, future studies should explore the value of incorporated sitting exercises into fall injury prevention programs, and assess whether an individual's risk for fall-related injury depends simply on whether they can land softly when descending into a sitting position.

Acknowledgements

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Appendix A. Definition of vertical kinetic energy

The kinetic energy of a system of links can be defined as

$$\mathrm{KE}_{\mathrm{tot}} = \sum_{i=1}^{n} \left[\tfrac{1}{2} m_i v_i^2 + \tfrac{1}{2} I_i \theta_i^2\right],$$

where n is the number of links, v_i is the translational velocity of the ith link's center of gravity, θ_i is the rotational velocity of the ith link, m_i is the mass of the ith link, $I_i = m_i \rho_i^2$ is the mass moment of inertia of the ith link, and ρ_i is the radius of gyration of the ith link about its center of gravity.

An alternate form of $\mathrm{KE}_{\mathrm{tot}}$ separates the translational velocity into vertical and horizontal components:

$$\mathrm{KE}_{\mathrm{tot}} = \sum_{i=1}^{n} \left[\tfrac{1}{2} m_i v_{\mathrm{h}_i}^2 + \tfrac{1}{2} m_i v_{\mathrm{v}_i}^2 + \tfrac{1}{2} I_i \theta_i^2\right]$$

or

$$\mathrm{KE}_{\mathrm{tot}} = \sum_{i=1}^{n} [\mathrm{KE}_{\mathrm{h}_i} + \mathrm{KE}_{\mathrm{v}_i} + \mathrm{KE}_{\mathrm{rot}_i}]$$

where the subscripts h, v, and rot represent horizontal, vertical, and rotational, respectively. Removing the summation notation gives $\mathrm{KE}_{\mathrm{tot}} = \mathrm{KE}_{\mathrm{h}} + \mathrm{KE}_{\mathrm{v}} + \mathrm{KE}_{\mathrm{rot}}$. In the current study, we used KE_{v} as an index of impact severity.

Appendix B. Definition of normalized vertical velocity and kinetic energy

During a fall from a static initial configuration, conservation of energy requires that

$$\mathrm{PE}_{\mathrm{i}} = \mathrm{PE}_{\mathrm{f}} + \mathrm{KE}_{\mathrm{f}} + W_{\mathrm{tot}}, \tag{B.1}$$

where PE_{i} is body's initial potential energy (before the initiation of decent), PE_{f} is the body's potential energy at impact, KE_{f} is the body's total kinetic energy at impact, and W_{tot} is the energy absorbed in the muscles, tendons and ligaments during descent. Let us consider a "worse-case" fall which involves $W_{\mathrm{tot}} = 0$, and model the body as a variable-length pendulum having an "effective mass" M situated at the hip. In this case, Eq. (B.1) becomes

$$Mgl = Mgb + \tfrac{1}{2}Mv^{p^2}, \tag{B.2}$$

where v^p is the translational velocity of the hip, and l and b are the vertical distances from the hip to the ankle at fall initiation and pelvis impact, respectively. Eq. (B.2) can then be rewritten as

$$v^p = \sqrt{2g(1-b)} = \sqrt{2gh}. \tag{B.3}$$

The vertical component of v^p is

$$v_v^p = v^p \cos\theta, \tag{B.4}$$

where

$$\cos\theta = \frac{a}{\sqrt{a^2+b^2}}, \tag{B.5}$$

and a is the horizontal distance from the hip to the ankle. Combining (B.3)–(B.5), the vertical velocity of the body during this worse-case fall is

$$v_{\mathrm{v}}^p = \sqrt{2gh}\,\frac{a}{\sqrt{a^2+b^2}}.$$

Based on similar considerations, the vertical kinetic energy of the body is

$$\mathrm{KE}_v^P = \frac{1}{2}M(v_{\mathrm{v}}^p)^2 = Mgh\left(\frac{a^2}{a^2+b^2}\right).$$

In the present study, attenuation of impact severity was expressed with the following normalized indices of vertical velocity and kinetic energy:

$$v_{\mathrm{v}}' = \frac{v_{\mathrm{v}}}{v_{\mathrm{v}}^p} = \frac{v_{\mathrm{v}}}{(\sqrt{2gh}\,a/\sqrt{a^2+b^2})} \tag{B.6}$$

and

$$\mathrm{KE}_{\mathrm{v}}' = \frac{\mathrm{KE}_{\mathrm{v}}}{\mathrm{KE}_{\mathrm{v}}^{\mathrm{p}}} = \frac{\mathrm{KE}_{\mathrm{v}}}{Mgh(a^2/(a^2+b^2))} \tag{B.7}$$

where v_{v} and KE_{v} are the vertical velocity and kinetic energy of the body during an actual trial.

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