



# Wrist impact velocities are smaller in forward falls than backward falls from standing

Juay-Seng Tan<sup>a</sup>, Janice J. Eng<sup>b,c,\*</sup>, Stephen N. Robinovitch<sup>d</sup>, Brady Warnick<sup>d</sup>

<sup>a</sup>*Division of Orthopaedic Engineering Research, University of British Columbia, Vancouver, BC, Canada*

<sup>b</sup>*School of Rehabilitation Sciences, University of British Columbia, Vancouver, BC, Canada*

<sup>c</sup>*Rehabilitation Research Laboratory, GF Strong Rehab Centre, Vancouver, BC, Canada*

<sup>d</sup>*School of Kinesiology, Simon Fraser University, Burnaby, BC, Canada*

Accepted 23 May 2005

---

## Abstract

The wrist is a common fracture site for both young and older adults. The purpose of this study was to compare wrist kinematics in backward and forward falls with different fall protective responses. We carried out within-subject comparison of impact velocities and maximum velocities during descent of the distal radius among three different fall configurations: (a) backward falls with knees flexed, (b) backward falls with knees extended and (c) forward falls with knees flexed. We also examined the effect of fall configuration on fall durations, elbow flexion, trunk flexion and forearm angles at impact. Forward falls resulted in smaller impact velocities of the distal radius, longer fall duration, longer braking duration, greater elbow flexion and more horizontal landing position of the forearm compared to backward falls. The distal radius impact velocity during forward falls (1.33 m/s) was significantly lower than in backward falls, and among the backward falls the impact velocity of the flexed knee strategy (2.01 m/s) was significantly lower than the extended knee strategy (2.27 m/s). These impact velocities were significantly reduced from the maximum velocities observed during descent (forward falls = 3.57 m/s, backward falls with knee flexed = 3.16 m/s, backward falls with knees extended = 3.52 m/s). We conclude that (1) smaller impact velocities of the wrists in forward falls could imply a lower fracture risk compared to backward falls, and (2) fall protective responses reduced wrist impact velocities in all fall directions.

© 2005 Elsevier Ltd. All rights reserved.

*Keywords:* Fall protective responses; Fall direction; Wrist fracture; Distal radius fracture; Impact velocity

---

## 1. Introduction

The distal radius is a common site of fracture (Adachi et al., 2001; Doherty et al., 2001; Melton et al., 1998; Sanders et al., 1999; Singer et al., 1998; Zebaze and Seeman, 2003). Distal radius fractures are particularly common in women with osteoporosis due to their compromised bone quality (Turner, 2002) and possibly due to the increased risk of falling in this population (Lui-Ambrose et al., 2003). In younger adults, injury to

the wrists are usually sports related, such as from in-line skating (Sherker and Cassell, 1999). The high rate of occurrence of a wrist fracture is evident from its lifetime risk of 13%, second only to hip fractures at 17% (Doherty et al., 2001). In fact, a higher risk of sustaining a wrist fracture than a hip fracture was observed for adults between 45 and 75 years old (Sanders et al., 1999; Singer et al., 1998). The direct medical cost in the US as a result of a wrist fracture was estimated to average USD\$1628 per patient in 1992 (Gabriel et al., 2002), and a total national cost for fall-related injuries of up to \$50 billion (in 1991 dollars) annually (DeGoede et al., 2003). Moreover, the physical, emotional and psychological effects resulting from osteoporotic fractures adversely influence health-related quality of life (Adachi et al.,

---

\*Corresponding author. School of Rehabilitation Sciences, University of British Columbia, Vancouver, BC, Canada. Tel.: +1 604 875 0615; fax.: +1 604 714 4168.

E-mail address: [janice.eng@vch.ca](mailto:janice.eng@vch.ca) (J.J. Eng).

2001). Patients often develop substantial pain, restricted motion and loss of function immediately following distal radius fracture.

Falls with a direct impact onto the extended hand are the cause of at least 90% of all wrist fractures (Nevitt et al., 1993; Palvanen et al., 2000). In older adults, forward falls occurred more often than falls in other directions (Nevitt et al., 1993). However, the risk of sustaining a wrist fracture is higher during backward falls (Nevitt et al., 1993). The mechanisms that result in this higher rate of fracture from backward falls are not fully understood. Biomechanical factors such as the body configuration and velocity at impact (DeGoede et al., 2003) may influence if a fracture of the wrist is to occur and these factors may vary depending on the direction of the fall.

The upper extremities are an important component of the fall protective response and the hands are commonly used to break the fall despite the risks of a wrist fracture. With appropriate fall protective responses, the forces to the wrist at impact could be reduced while attaining a safe body configuration at impact (DeGoede et al., 2003).

Studies on fall protective responses have focused on forward falls (Chou et al., 2001; DeGoede and Ashton-Miller, 2002, 2003; DeGoede et al., 2002; Kim and Ashton-Miller, 2003) with fewer studies in the backward direction (Hsiao and Robinovitch, 1998; Sandler and Robinovitch, 2001). No direct comparisons between these two directions exist. Given the higher frequency of sustaining a wrist fracture during backward falls (Nevitt

et al., 1993), studies which lead to a better understanding of the effects of fall protective responses in this direction are necessary.

The purpose of this study was to compare maximum and impact velocities of the wrist during backward and forward falls in young adults. Given the reported increased risk of fracture with a backward fall, we hypothesized a higher impact velocity of the wrist in the backward direction compared to the forward direction during a fall.

## 2. Methods

Fifteen healthy young female subjects were recruited for this study. Their ages ranged from 19 to 30 years (mean = 22, s.d. = 3) and all were right hand dominant. Prior training in fall strategies such as in martial arts was an exclusion criterion. Following university ethics approval, informed consent was received from all participants prior to their participation.

Subjects stood with hands by their sides and were released from an unbalanced standing position and fell onto a gymnasium mattress (Fig. 1). For the forward falls, subjects stood with a 15° forward lean supported by a tether, and for the backward falls, subjects stood with a 5° backward lean. These angles were determined from previous research to represent values that just exceed the capacity of young subjects to recover from falling by taking a single step (Hsiao and Robinovitch, 1999, 2001). The tether was released via an electro-

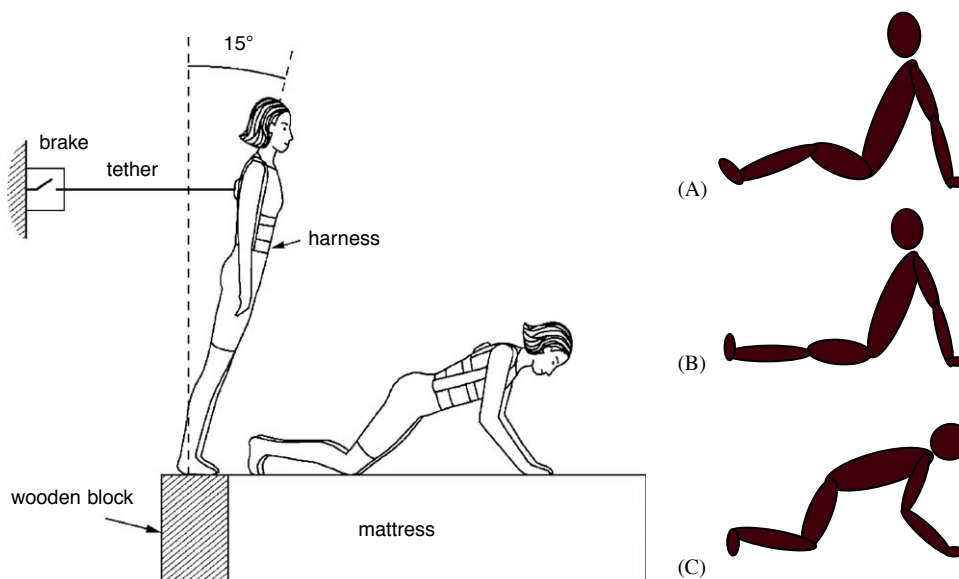


Fig. 1. Subjects stood supported at an angle of 15° to the vertical in forward falls and 5° in backward falls. A tether was released causing them to fall unexpectedly, landing simultaneously on their buttocks and hands in backward falls or knees first followed by hands in forward falls. Buttocks impact in backward falls as a proportion of the fall duration occurred at  $t = 99\%$  (s.d. = 5%) with knees flexed and at  $t = 98\%$  (s.d. = 4%) with knees extended. Knees impact in forward falls occurred significantly earlier at  $t = 77\%$  (s.d. = 6%) of the fall duration. The schematics on the right show body configurations at impact for (A) backward falls with knees flexed, (B) backward falls with knee extended, and (C) forward falls with knees flexed.

magnetic catch with a delay of between 1 and 10 s after the verbal cue ('ready?') to increase the unexpectedness of the fall. Subjects were instructed to fall in three different configurations: (a) backward fall with knees flexed, (b) backward fall with knees extended and (c) forward fall with knees flexed (Fig. 1). Three trials were collected for each configuration. The sequence of conducting the three fall configurations was randomised, with the three trials for each configuration conducted in sequence. In backward falls, subjects impacted the hands and buttocks simultaneously, while in forward falls, subjects landed on their knees first followed by the hands (Fig. 1). Subjects fell symmetrically without taking a step forward or backward. For all fall configurations, subjects were instructed to (1) "land as softly as possible", (2) land with both hands simultaneously and (3) prevent impact to the head (forward falls) or reduce impact to the hips (backward falls). The subjects utilised fall protective responses such as elbow flexion in all falls.

A seven-camera motion measurement system (ProReflex, Qualysis Inc., East Windsor, CT) was used to measure the position of 22 spherical reflective markers attached to the top of the head, acromion, distal humerus (lateral epicondyle), distal radius, sacrum, anterior–superior iliac spine (ASIS), proximal femur (greater trochanter), distal femur (lateral epicondyle), anterior tibia (mid-shaft), distal fibula (lateral malleolus), third metatarsal and calcaneus. Three additional markers placed on the surface of the mattress were used to define the impact surface. Motion data were captured at 60 Hz.

Kinematic data were analysed using custom routines (MATLAB, MathWorks, Natick, MA and Microsoft Excel) to determine (a) the maximum and (b) impact velocities of the wrist, (c) time durations (as defined below), (d) shoulder to wrist (shoulder–wrist) distances at impact, (e) shoulder to hip (shoulder–hip) distances at impact and (f) forearm angles at impact. Impact velocity is proportional to impact force and could be feasibly and ethically measured rather than impact force. Three distal radius velocity measures were analysed: horizontal ( $|V_y|$ ), vertical ( $|V_z|$ ) and resultant ( $|V|$ , where  $\vec{V} = \vec{V}_y + \vec{V}_z$ ) velocities. Motion and velocity in the medial–lateral direction ( $x$ -axis) during falls were small compared to sagittal motions and were not included in these analyses. Maximum velocities were determined from the velocity–time plots (Fig. 2) and occurred prior to impact in all trials. The instance of impact was defined as the time when the distal radius marker crossed the horizontal plane of the support surface and impact velocity was the velocity of the distal radius at the instance of impact. An independent observer repeated the kinematic analysis to determine the instance of impact and the mean difference between observers was 0.3 timestep (s.d. = 0.7) or 0.005 s

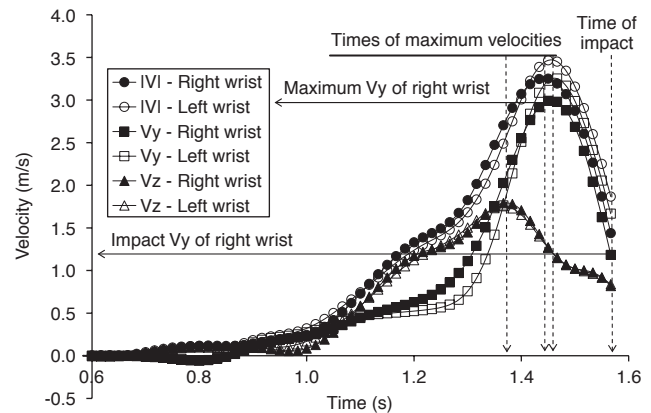


Fig. 2. Velocity–time profiles of both wrists for a subject during a single forward fall. The horizontal ( $|V_y|$ ), vertical ( $|V_z|$ ) and resultant ( $|V|$ ) velocities of both wrists were analysed. Impact times were determined as the instance when the distal radius markers crossed the horizontal plane of the support surface. Impact velocities were subsequently determined from velocity–time data. Velocity profiles of the left and right wrists were generally similar.

(s.d. = 0.01). Shorter shoulder–wrist distances indicated elbow flexion and shorter shoulder–hip distances indicated trunk flexion. The forearm angle was calculated as an acute angle in the sagittal plane determined by the wrist and elbow markers, relative to the horizontal plane. Fall duration was the time from fall initiation to time of hands impact and was divided into an initiation phase (initiation to maximum resultant velocity) and a braking phase (maximum resultant velocity to impact) (Fig. 3). The impact time of the buttocks and knees was determined as a proportion of the fall duration.

A reduction in the velocity from the time at which maximum velocity occurs to the time of impact was indicative of the successful application of fall protective responses. Differences in fall duration (initiation and braking durations) between fall directions were indicative of different fall protective responses. Lastly, forearm angles at impact were relevant to the impact forces and moments experienced by the radius and its risk of fracture.

The final analyses included 102 trials from 15 subjects with data bilaterally and a further 19 trials with data unilaterally, giving an overall 83% inclusion from all 135 possible trials. There were no data for seven trials due to technical issues in data collection. Secondly, there were three bilateral and 12 unilateral occlusions of the distal radius markers from camera view at the instance of impact. A further 11 trials (four bilaterally and seven unilaterally) were excluded as they were identified as outliers from coefficients of variation (COV) analyses. Impact velocities of these 11 trials were greater than two standard deviations from the subject's mean and were excluded from further analyses. Mean within-subject

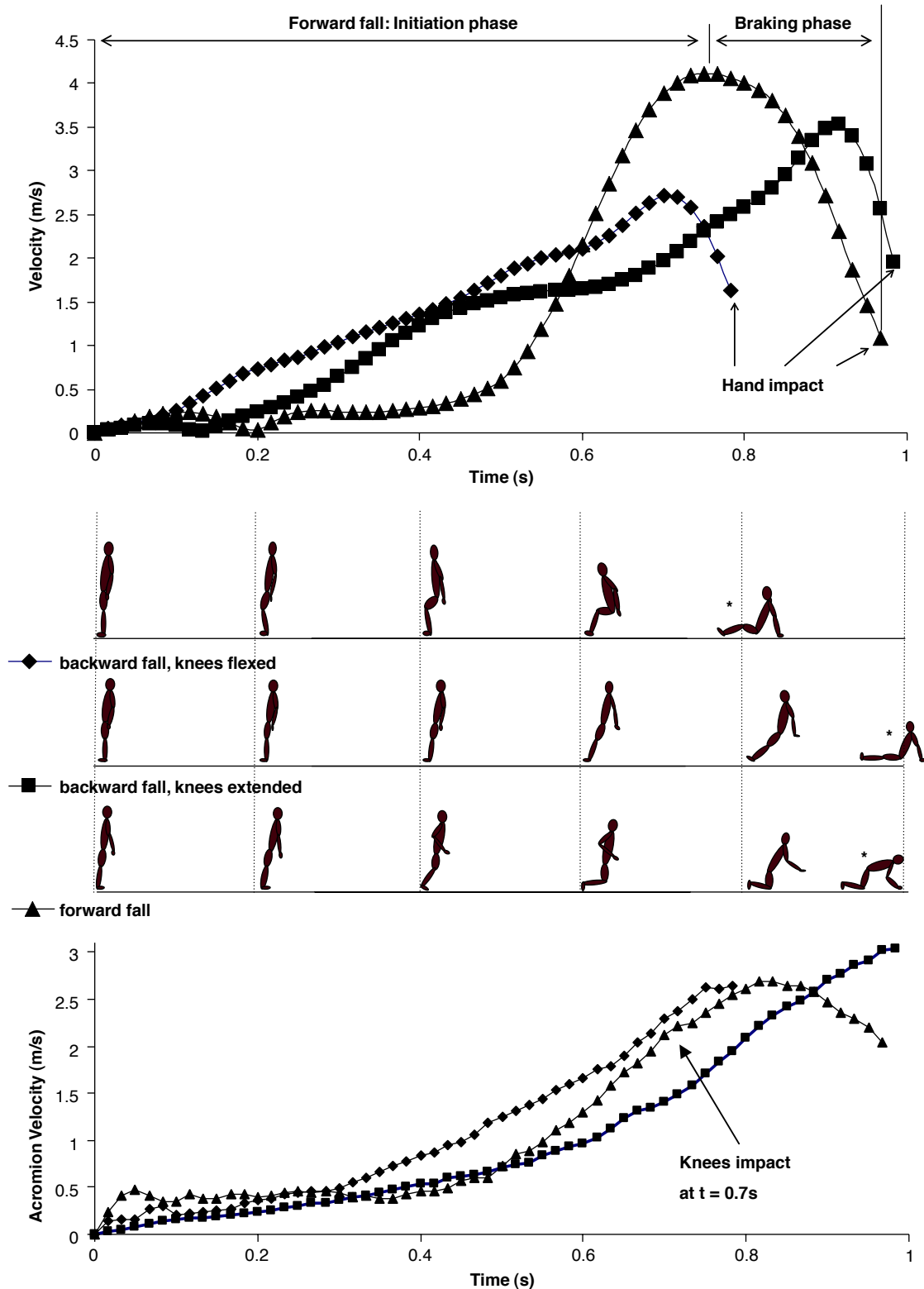


Fig. 3. Velocity–time profiles ( $|V|$ ) of the right wrist of a single subject in three different fall configurations from initiation to impact. Initiation phase was defined as the time between fall initiation to maximum velocity and braking phase was defined as the time between maximum velocity and impact. Fall duration was the sum of the initiation and braking phases. Initiation phase lasts about 0.8 s while braking phase lasts about 0.2 s. For forward falls, fall durations and braking phase were significantly longer while impact velocity was significantly smaller. The plots below present the velocity–time profiles ( $|V|$ ) of the right acromion marker for the same trial. A distinctive decrease in acromion velocity was observed for forward falls, after knee impact at  $t = 0.7$  s.

COV for  $|V_y|$ ,  $|V_z|$  and  $|V|$  for the remaining accepted trials were 24%, 20% and 17%, respectively.

We conducted paired *t*-tests with a 0.05  $\alpha$  level between dominant and non-dominant hands on (1) maximum and (2) impact velocities, for each fall configuration. In cases where there were side-to-side differences found with the *t*-tests, we used the average value of the three trials from the hand with the higher velocity for further analyses to represent a worse-case scenario for fracture risk. For velocities that were not different between hands, the six data values (three trials, two hands) were averaged by subject. Significant differences between the dominant and non-dominant hand velocities were found in some fall configurations and velocity measures (Table 1). No significant differences in impact forearm angles between dominant and non-dominant hands for all fall configurations were found and the average values from both hands and three trials were presented.

Three separate two-way ANOVAs for  $|V_y|$ ,  $|V_z|$  and  $|V|$  blocked for subjects were used to compare the velocity event (maximum vs. impact velocity) (independent factor 1) among the three fall configurations (independent factor 2). Three one-way ANOVAs blocked for subjects were used to compare time durations: (1) fall duration, (2) initiation phase and (3) braking phase among the three fall configurations (independent factor). Next, a two-way ANOVA blocked for subjects was used to compare the joint angles (shoulder–wrist and shoulder–hip distances) (independent factor 1) among the three fall configurations (independent factor 2). Lastly, a one-way ANOVA blocked for subjects was used to compare forearm impact angle for the three fall configurations (independent

factor). Duncan post hoc tests were used following significant ANOVA results. All tests were performed using statistical analysis software (Statistica, Statsoft Inc., Tulsa, OK), with an  $\alpha$  of 0.05.

### 3. Results

There was a significant two-way interaction (velocity event  $\times$  fall configuration) and significant main effect of velocity event (impact vs. maximum velocity) for the horizontal  $|V_y|$  and resultant  $|V|$  (Table 1). For the vertical  $|V_z|$ , there was a significant main effect of velocity event and significant main effect of fall configuration. The resultant  $|V|$  impact velocities were significantly reduced by 63%, 36% and 36%, from resultant  $|V|$  maximum velocities for fall configurations A, B and C, respectively (Fig. 1). Post-hoc tests showed that (1) all impact velocities ( $|V_y|$ ,  $|V_z|$ , and  $|V|$ ) were significantly smaller in forward falls compared to backward falls and (2) the vertical  $|V_z|$  and resultant  $|V|$  impact velocities were significantly higher in backward falls with knees extended compared to the other two fall configurations. In contrast, the maximum horizontal velocity  $|V_y|$  was significantly higher in forward falls compared to the other fall configurations, however, maximum vertical velocity  $|V_z|$  was greatest during the backward falls with knees extended compared to the other two fall configurations.

There was a significant effect of fall configuration on fall duration. Post-hoc tests revealed that fall durations for backward falls with knees flexed (0.749 s) were shorter ( $p = 0.0001$ ) than the other fall configurations, while the forward falls (0.980 s) were longer ( $p = 0.0001$ )

Table 1  
Kinematic parameters among the three fall configurations

		(a) Backward fall, knees flexed	(b) Backward fall, knees extended	(c) Forward fall
Wrist impact velocity (m/s)	Horizontal, $ V_y $	1.24 <sup>a</sup>	1.25 <sup>a</sup>	0.64 <sup>d</sup>
	Vertical, $ V_z $	1.59 <sup>b</sup>	1.89 <sup>c</sup>	1.14 <sup>d</sup>
	Resultant, $ V $	2.01	2.27 <sup>c</sup>	1.33 <sup>d</sup>
Wrist maximum velocity (m/s)	Horizontal, $ V_y $	2.06 <sup>c</sup>	2.05 <sup>c</sup>	2.79 <sup>a,c,e</sup>
	Vertical, $ V_z $	2.34 <sup>c</sup>	2.83 <sup>c,e</sup>	2.13 <sup>c</sup>
	Resultant, $ V $	3.16 <sup>c,d</sup>	3.52 <sup>c</sup>	3.57 <sup>c</sup>
Time durations (s)	Fall duration	0.749 <sup>d</sup>	0.873	0.980 <sup>e</sup>
	Initiation phase	0.665 <sup>d</sup>	0.799	0.787
	Braking phase	0.086	0.071	0.190 <sup>c</sup>
Distances at impact (mm)	Shoulder–wrist	488	485	441 <sup>d</sup>
	Shoulder–hip	412 <sup>d</sup>	438	548 <sup>c</sup>
Forearm impact angle (deg)	Sagittal plane	70	69	45 <sup>d</sup>

<sup>a</sup>Significantly higher velocity in the dominant hand.

<sup>b</sup>Significantly higher velocity in the non-dominant hand.

<sup>c</sup>Significantly higher than corresponding impact velocity,  $p < 0.05$ .

<sup>d</sup>Significantly lower than the other two fall configurations,  $p < 0.05$ .

<sup>e</sup>Significantly higher than the other two fall configurations,  $p < 0.05$ .

than the other fall configurations (Table 1 and Fig. 3). Initiation phase in all falls lasted between means of 0.665 s (backward falls with knees flexed) and 0.799 s (backward falls with knees extended), with maximum velocity occurring sooner in backward falls with knee flexed ( $p = 0.0001$ ) compared to the other two configurations. Lastly, the braking duration for forward falls (0.190 s) was significantly longer ( $p = 0.0001$ ) compared to the other two fall configurations. Resultant velocities of the distal radius peaked at 89% and 92% of the fall duration in backward falls with knee flexed and knee extended, respectively, while velocities of the distal radius peaked at 80% of the fall duration in forward falls. During backward falls with knees flexed, the buttocks contacted the ground at 99% (s.d. = 5%) of the fall duration and at 98% (s.d. = 4%) when the knees were extended. Knee impact during forward falls occurred significantly earlier at 77% (s.d. = 6%) of the fall duration.

Two-way ANOVA on shoulder–wrist and shoulder–hip distances among the three fall configurations found a significant two-way interaction (joint angle  $\times$  fall configuration) ( $p < 0.0001$ ) and significant main effect of fall configuration ( $p < 0.0001$ ) (Table 1 and Fig. 4). Post-hoc analysis showed that (1) shoulder–wrist distance at impact was significantly shorter in forward falls (indicating more elbow flexion) than backward falls, (2) shoulder–hip distance was significantly larger in forward falls (indicating less flexion of the torso) than in backward falls, and (3) shoulder–hip distance was significantly smaller in backward falls with knees flexed compared to the other two fall configurations.

There was a significant effect of fall configuration on forearm impact angles. The position of the forearms at

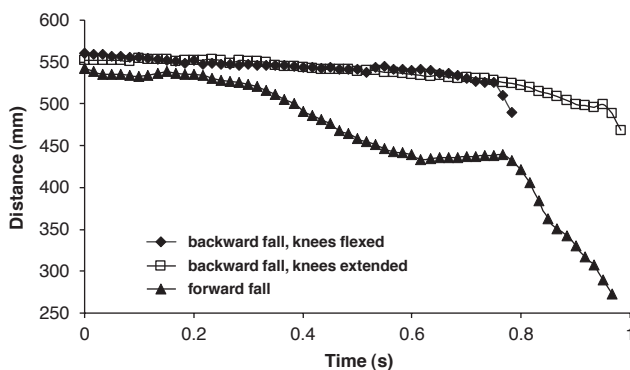


Fig. 4. Absolute distance between shoulder and wrist markers (shoulder–wrist) of the right arm of a single subject (same subject as Fig. 3) in three different fall configurations (similar trials as Fig. 3). Shorter shoulder–wrist distance indicates elbow flexion. Initial distance was about 550 mm apart for all three fall configurations. During the fall, shoulder–wrist distance decreases regardless of fall configuration but with higher decreases in forward falls, especially during the braking phase (last 0.2 s) and after knees impact to ground. At impact, shoulder–wrist distance was significantly shorter for forward falls, indicating more elbow flexion.

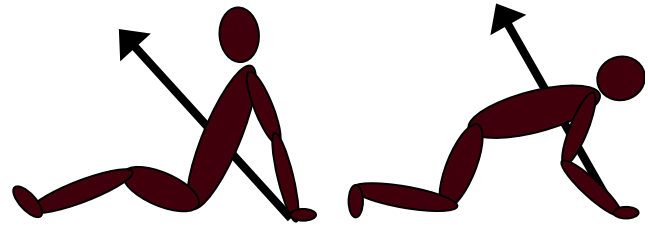


Fig. 5. Resultant ground reaction force directions, assuming impact force and impact velocity vectors are in the same direction. Force vectors resulted in a flexion moment on the forearm in backward falls and an extension moment in forward falls. The force vectors were  $52^\circ$  and  $61^\circ$  to the horizontal plane in backward and forward falls respectively, while the corresponding forearm angles were  $70^\circ$  and  $45^\circ$ .

impact for backward falls was significantly more vertical at  $70^\circ$  (with knees flexed) and  $69^\circ$  (with knees extended) ( $p = 0.0002$ ), compared to forward falls ( $45^\circ$ ) (Table 1, Fig. 5).

#### 4. Discussion

The purpose of this study was to compare impact velocities of the distal radius between backward and forward falls. Fall protective responses utilised by subjects in this study included (1) the simultaneous impact to the pelvis and wrists in backward falls, (2) knee contact prior to wrist impact in forward falls, (3) elbow flexion in both backward and forward falls, and (4) trunk flexion in backward falls.

Forward falls resulted in smaller impact velocities to the distal radius. The resultant wrist impact velocity during forward falls was only 66% and 59% of the values found in backward falls with knees flexed and extended, respectively. This reflected the protective role (in terms of risk for upper extremity injury) of the initial impact to the knees during forward falls, which absorbed energy and thereby lowered the subsequent velocities of the upper body and hands. Since impact velocity is directly related to impact force, smaller impact velocities in forward falls should correspond to lower impact forces on the distal radius. This may contribute to the observed greater risk for wrist fracture for a backward fall compared to a forward fall (Nevitt et al., 1993).

We also found that impact velocities were higher in backward falls with knees extended than for the other two configurations, reflecting the protective benefit of absorbing energy in the lower extremity joints during descent (Hsiao and Robinovitch, 1998; Sandler and Robinovitch, 2001). A recent study showed that the effectiveness of this “squat protective response” depends on the stage during descent when it is initiated, diminishing in benefit as the fall progresses and the

state of imbalance grows increasingly severe (Robinovitch et al., 2004).

Lower extremities dynamics (knee and corresponding hip flexion) in addition to upper extremities dynamics (elbow and trunk flexion) contributed to the braking phase during the fall. The elbow flexion observed in this study contributed to the reduction in velocity during forward falls and is consistent with previous studies (Chou et al., 2001; DeGoede and Ashton-Miller, 2002, 2003; DeGoede et al., 2002; Kim and Ashton-Miller, 2003). Elbow flexion during forward falls was also shown to influence the force profile after impact and reduce fracture risk. The body segments which contacted the ground prior to the wrists will also absorb impact energy and thereby influence the wrist impact velocity; the braking phase in forward falls was prolonged by an early knee impact, while in backward falls simultaneous impact of the buttocks and hands occurred. The velocity profile of the acromion markers (Fig. 3) indicated a reduction in upper body velocities after knee impact during the forward fall. Greater visual feedback of the hands and of the impact surface during forward falls could also have contributed to the longer braking phase and smaller impact velocities.

In backward falls, fall protective responses from both upper and lower extremities dynamics were also observed. Knee flexion resulted in smaller maximum and impact velocities of the distal radius than with extended knees. This observation is consistent with an earlier study (Sandler and Robinovitch, 2001) where knee flexion and energy absorption lowered vertical impact velocities at the pelvis by a mean of 28% (s.d. = 15). Thus knee flexion was an effective fall protective response during backward falls. Elbow flexion was also one of the strategies exhibited by the subjects during backward falls. Elbow flexion during backward falls was however limited by the vertical clearance between the wrists and buttocks at impact. In order for the hands to touch the ground before the pelvis, the vertical shoulder–wrist distance had to be larger than the vertical shoulder–hip distance. To increase elbow flexion in backward falls, subjects were observed to flex their trunk during backward falls.

The forearm angle at impact was different between backward and forward falls, with a more vertical forearm during backward falls. The influence of the resultant force direction on fracture risk is a variable that is not well understood. Differences in forearm position and velocity vector (and presumably force vector) that we measured would result in an extension moment in forward falls and a flexion moment on the forearm in backward falls (Fig. 5).

Fracture risks were shown in previous studies to be influenced by parameters such as impact velocity (DeGoede and Ashton-Miller, 2002; DeGoede et al., 2002; van den Kroonenberg et al., 1996) and elbow

angle (Chou et al., 2001; DeGoede and Ashton-Miller, 2002; DeGoede et al., 2002). Other biomechanical variables that could influence fracture risk but were not measured in the current study include (a) impact force (Chiu and Robinovitch, 1998), (b) wrist flexion/extension and (c) forearm pronation/supination. Moreover, failure force of the distal radius may differ between different impact configurations due to differences in angular acceleration of the forearm and body configuration.

In conclusion, forward falls resulted in smaller impact velocities of the distal radius than backward falls when fall protective responses were applied and this could translate to a lower fracture risk during forward falls. Fall protective responses during both forward and backward falls reduced impact velocities of the distal radius from the maximum velocities.

### Acknowledgments

We would like to acknowledge career investigator awards from the Canadian Institute for Health Research (CIHR) and the Michael Smith Foundation for Health Research to J. Eng and S. Robinovitch. We also acknowledge operating grants from NIH (R01AR46890) and CDC (R49/CCR019335).

### References

- Adachi, J.D., Ioannidis, G., Berger, C., Joseph, L., et al., 2001. The influence of osteoporotic fractures on health-related quality of life in community-dwelling men and women across Canada. *Osteoporosis International* 12, 903–908.
- Chiu, J., Robinovitch, S.N., 1998. Prediction of upper extremity impact forces during falls on the outstretched hand. *Journal of Biomechanics* 31, 1169–1176.
- Chou, P.H., Chou, Y.L., Lin, C.J., Su, F.C., et al., 2001. Effect of elbow flexion on upper extremity impact forces during a fall. *Clinical Biomechanics* 16, 888–894.
- DeGoede, K.M., Ashton-Miller, J.A., 2002. Fall arrest strategy affects peak hand impact force in a forward fall. *Journal of Biomechanics* 35, 843–848.
- DeGoede, K.M., Ashton-Miller, J.A., Schultz, A.B., Alexander, N.B., 2002. Biomechanical factors affecting the peak hand reaction force during the bimanual arrest of a moving mass. *Journal of Biomechanical Engineering* 124, 107–112.
- DeGoede, K.M., Ashton-Miller, J.A., 2003. Biomechanical simulations of forward fall arrests: effects of upper extremity arrest strategy, gender and aging-related declines in muscle strength. *Journal of Biomechanics* 36, 413–420.
- DeGoede, K.M., Ashton-Miller, J.A., Schultz, A.B., 2003. Fall-related upper body injuries in the older adult: a review of the biomechanical issues. *Journal of Biomechanics* 36, 1043–1053.
- Doherty, D.A., Sanders, K.M., Kotowicz, M.A., Prince, R.L., 2001. Lifetime and five-year age-specific risks of first and subsequent osteoporotic fractures in postmenopausal women. *Osteoporosis International* 12, 16–23.

- Gabriel, S.E., Tosteson, A.N.A., Leibson, C.L., Crowson, C.S., et al., 2002. Direct medical costs attributable to osteoporotic fractures. *Osteoporosis International* 13, 323–330.
- Hsiao, E.T., Robinovitch, S.N., 1998. Common protective movements govern unexpected falls from standing height. *Journal of Biomechanics* 31, 1–9.
- Hsiao, E.T., Robinovitch, S.N., 1999. Biomechanical influences on balance recovery by stepping. *Journal of Biomechanics* 32, 1099–1106.
- Hsiao, E.T., Robinovitch, S.N., 2001. Elderly subjects' ability to recover balance with a single backward step associates with body configuration at step contact. *The Journals of Gerontology, Series A, Biological Sciences and Medical Sciences* 56, M42–47.
- Kim, K.J., Ashton-Miller, J.A., 2003. Biomechanics of fall arrest using the upper extremity: age differences. *Clinical Biomechanics* 18, 311–318.
- Lui-Ambrose, T., Eng, J.J., Khan, K.M., Carter, N.D., McKay, H.A., 2003. Older women with osteoporosis have increased postural sway and weaker quadriceps strength than counterparts with normal bone mass: overlooked determinants of fracture risk? *Journal of Gerontology* 58, M862–M866.
- Melton III., L.J., Amadio, P.C., Crowson, C.S., O'Fallon, W.M., 1998. Long-term trends in the incidence of distal forearm fractures. *Osteoporosis International* 8, 341–348.
- Nevitt, M.C., Cummings, S.R., et al., 1993. Types of fall and risk of hip and wrist fractures: the study of osteoporotic fractures. *Journal of the American Geriatrics Society* 41, 1226–1234.
- Palvanen, M., Kannus, P., Parkkari, J., Pitkajarvi, T., et al., 2000. The injury mechanisms of osteoporotic upper extremity fractures among older adults: a controlled study of 287 consecutive patients and their 108 controls. *Osteoporosis International* 11, 822–831.
- Robinovitch, S.N., Brumer, R., Maurer, J., 2004. Effect of the “squat protective response” on impact velocity during backward falls. *Journal of Biomechanics* 37, 1329–1337.
- Sanders, K.M., Seeman, E., Ugoni, A.M., Pasco, J.A., et al., 1999. Age- and gender-specific rate of fractures in Australia: a population based study. *Osteoporosis International* 10, 240–247.
- Sandler, R., Robinovitch, R., 2001. An analysis of the effect of lower extremity strength on impact severity during a backward fall. *Journal of Biomechanical Engineering* 12, 590–598.
- Sherker, S., Cassell, E., 1999. Preventing in-line skating injuries. How effective are the countermeasures? *Sports Medicine* 28, 325–335.
- Singer, B.R., McLauchlan, G.J., Robinson, C.M., Christie, J., 1998. Epidemiology of fractures in 15000 adults. *The Journal of Bone and Joint Surgeries (Br)* 80-B, 243–248.
- Turner, C.H., 2002. Biomechanics of bone: determinants of skeletal fragility and bone quality. *Osteoporosis International* 13, 97–104.
- van den Kroonenberg, A.J., Hayes, W.C., McMahon, T.A., 1996. Hip impact velocities and body configurations for voluntary falls from standing height. *Journal of Biomechanics* 29, 807–811.
- Zebaze, R.M.D., Seeman, E., 2003. Epidemiology of hip and wrist fractures in Cameroon, Africa. *Osteoporosis International* 14, 301–305.