HIP IMPACT VELOCITIES AND BODY CONFIGURATIONS FOR VOLUNTARY FALLS FROM STANDING HEIGHT

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Abstract—Fall dynamics have largely been ignored in the study of hip fracture etiology and in the development of hip fracture prevention strategies. In this study, we asked the following questions: (1) What are the ranges of hip impact velocities associated with a sideways fall from standing height? (2) What are the ranges of body configurations at impact? and (3) How do protective reflexes such as muscle activation or using an outstretched hand influence fall kinematics? To answer these questions, we recruited six young healthy athletes who performed voluntary sideways falls on a thick foam mattress. Several categories of falls were investigated: (a) muscle-active vs muscle-relaxed falls; (b) falls from a standing position or from walking; and (c) falls in which an outstretched arm was used to break the fall. Each fall was videotaped at 60 frames s⁻¹. Fall kinematics parameters were obtained by digitizing markers placed on anatomical points of interest. The mean value for vertical hip impact velocity was 2.75 m s⁻¹ (± 0.42 m s⁻¹ [S.D.]). The mean value for trunk angle (the angle between the trunk and the vertical) was 17.3° (± 11.5° [S.D.]). We found a 38% reduction in the trunk angle at impact, and a 7% reduction in hip impact velocity for relaxed vs muscle-active falls. Finally, regarding the falls in which an outstretched arm was used, only two out of the six subjects were able to break the fall with their arm or hand. For the remaining subjects, hip impact occurred first, followed by contact of the arm or hand. Published by Elsevier Science Ltd.

Keywords: Falls; Hip fractures; Kinematics; Aging.

INTRODUCTION

Falls are the leading cause of death from injury among persons aged 65 years and older (Rice et al., 1989). Major morbidity from falls includes more than 280,000 hip fractures per year (American Academy of Orthopaedic Surgeons, 1993). Osteoporosis is widely viewed as the primary cause of the increase in hip fracture incidence among the elderly. However, densitometric studies have shown significant overlap in bone density between hip fracture patients and age- and gender-matched controls (Greenspan et al., 1994). Since over 90% of all hip fractures occurring in the United States are caused by falls (Cummings et al., 1990), the increased incidence of falls among the elderly may be an important contributor to the dramatic increase in hip fracture risk for this age group. However, the increase in risk of a hip fracture for elderly persons compared to a middle-age population is more than ten times larger than the increase in the incidence of falls in the elderly (Fitti, 1987; Gallagher et al., 1980). Cummings et al. (1989) proposed that besides osteoporosis and falling, hip fractures are the result of several age-related changes in neuromuscular function that increase the severity of a fall, and therefore the likelihood that a fall will result in a hip fracture.

To explore this last hypothesis, we need to have a better understanding of fall dynamics and their relation to hip fracture. From a surveillance study among nursing home fallers, Greenspan et al. (1994) reported that the greatest risk factor for fracture was falling to the side, followed by a factor measuring bone mineral density and other factors measuring body habitus and the potential energy available in the fall. The authors concluded that in addition to bone mineral density, factors related to the dynamics of the fall may well dominate the risk of fracture from a fall. To date, however, hip fracture prevention efforts have focused primarily on reduction of bone loss (Raisz et al., 1993) and, if directed towards falling, on fall prevention (Tideiksaar, 1993; Tinetti et al., 1994). Few studies have approached hip fracture prevention through attempts to reduce the injury potential of those falls that do occur.

In order to develop new intervention strategies such as protective trochanteric padding or impact-attenuating floors, data are needed on the kinematics and dynamics of falls. Such data are also necessary for testing and improving analytical models of falling (Kroonenberg et al., 1995). The most valuable evidence concerning human falls would be derived from accidental, unexpected falls on hard floors. Safety and ethical considerations limit the possibilities for obtaining such information. In the study reported here, young athletes fell on a 0.6-m thick foam mattress of the type used for the landing pit of a high jump. We attached particular significance to two variables describing the body at impact: the vertical velocity of the hip and the angle of the trunk with respect to the vertical. We have argued elsewhere that the peak impact force would be expected to increase if either the impact velocity or the effective mass moving downward with the hip increases (Kroonenberg et al., 1995). Models of human falls based on rigid links coupled by hinges predict an apparent mass of the hip that increases if the trunk is made more vertical.
Therefore, in this study we addressed the following questions: (1) What are the ranges of hip impact velocities associated with a sideways fall from standing height? (2) What are the ranges of body configuration at impact, as measured by the angle of the trunk with the vertical? (3) How do protective reflexes such as muscle activation or the use of an outstretched hand influence fall kinematics?

MATERIALS AND METHODS

Six subjects ranging in age from 19 to 30 y (mean: 23.7 y ± 3.67 [S.D.]) and in height and weight from 1.63 to 1.93 m (mean: 1.76 m ± 0.12 [S.D.]) and from 534 to 801 N (mean: 651 N ± 116 [S.D.]), respectively, participated in the study. The falls were initiated by the subjects, based on instructions to ‘launch’ themselves (by introducing an initial sideways velocity of the trunk) and subsequently to fall as naturally as possible. Falls from two different initial body configurations were performed. In the first configuration, subjects were standing next to the mattress at the end of a wooden elevated walkway (Fig. 1). Subjects then slowly lifted their left foot before falling to the left. In the second configuration, they walked slowly on the walkway along the edge of the mattress before falling. We used a thick landing mattress, with the height of its surface about 0.12 m above the elevated platform (Fig. 1). To investigate the effect of muscle activity on fall dynamics, the subjects were instructed either to fall as relaxed as they could, almost as if they had fainted, or, in another series, to fall naturally, using the musculature of their lower extremity as they would in a ‘normal’ reflex-mediated fall. Also, in some cases the subjects were instructed to use their left arm to break the fall. Each subject performed between 2 and 6 trials in each fall category, depending on factors such as fatigue and the level of physical comfort of the subjects during the experiments. The fall categories were: W-AC (walking, muscle-active fall); ST-AC (standing, muscle-active fall); W-RX (walking, muscle-relaxed fall); ST-RX (standing, muscle-relaxed fall); W-ARM (walking, fall with outstretched arm) and ST-ARM (standing, fall with outstretched arm). The study was approved by the appropriate Institutional Review Board (Committee on Use of Human Subjects in Teaching and Research) at Harvard University. Each subject signed a consent form before participating in the experiments.

Surface electromyography (EMG) was used as a qualitative indicator of the muscular activity of the quadriceps (vastus lateralis) and of the hamstrings (medial head of the biceps femoris). This was particularly useful during the ‘practice sessions’ preceding the actual experiments, to teach the subjects how to fall relaxed. We recorded EMG activity in muscles acting to extend the knee because only these muscles, among all the muscles of the body, are capable of absorbing energy as the knee flexes, thereby reducing the impact velocity of the hip in a fall. The falls were video-taped at 60 frames s⁻¹ and displayed in real-time on a computer screen in combination with traces of both EMG signals and the vertical component of the foot force (Fig. 1). In order to estimate the hip impact velocities, the trajectory of a marker placed on the skin over the left greater trochanter was digitized. If the marker disappeared from view due to rotation of the pelvis, an outline of the body on a transparent sheet was used to estimate the location of the marker. Fourth-order polynomials were fit to the resulting y-displacement vs time curves. Next, the functions obtained were differentiated with respect to time and the hip impact velocities were evaluated at the time just before impact on the mattress. We also estimated the velocities that would have occurred had the mattress not been present and the subjects had impacted the floor. First, using a Newton–Raphson iteration scheme, we extrapolated the y vs time curve to estimate the time at which impact with the floor would have occurred. Then, the hip impact velocities were estimated by evaluating the differentiated polynomials at these new time points. To validate this procedure, the same technique was used to estimate the hip velocity that occurred at the mattress for a typical fall. Assuming that the last 0.12 m of y-displacements of the hip above the mattress surface were unknown, their vertical hip impact velocity at the mattress was estimated using extrapolation, according to the method described above. The percentage difference between the extrapolated and ‘real’ value for the hip impact velocity at the mattress is expected to be the same when comparing the extrapolated and ‘real’ value for the hip impact velocity at the floor level. The body configuration at impact was...
Table 1. Mean values \([ \pm \text{S.D.} ]\) of hip impact velocities at mattress (m/s)

<table>
<thead>
<tr>
<th>Subject</th>
<th>ST-RX (\pm)</th>
<th>ST-AC (\pm)</th>
<th>ST-ARM (\pm)</th>
<th>W-RX (\pm)</th>
<th>W-AC (\pm)</th>
<th>W-ARM (\pm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>45</td>
<td>2.96 (\pm) 0.25</td>
<td>3.17 (\pm) 0.15</td>
<td>3.12 (\pm) 0.13</td>
<td>2.57 (\pm) 0.15</td>
<td>2.98 (\pm) 0.04</td>
<td>3.49 (\pm) 0.17</td>
</tr>
<tr>
<td>67</td>
<td>2.84 (\pm) 0.13</td>
<td>2.91 (\pm) 0.12</td>
<td>*</td>
<td>3.26 (\pm) 0.09</td>
<td>3.61 (\pm) 0.65</td>
<td>*</td>
</tr>
<tr>
<td>89</td>
<td>2.23 (\pm) 0.20</td>
<td>2.25 (\pm) 0.15</td>
<td>2.17 (\pm) 0.10</td>
<td>2.13 (\pm) 0.24</td>
<td>2.34 (\pm) 0.05</td>
<td>2.05 (\pm) 0.11</td>
</tr>
<tr>
<td>10</td>
<td>2.64 (\pm) 0.30</td>
<td>2.54 (\pm) 0.11</td>
<td>*</td>
<td>2.52 (\pm) 0.54</td>
<td>2.57 (\pm) 0.11</td>
<td>*</td>
</tr>
<tr>
<td>11</td>
<td>2.79 (\pm) 0.25</td>
<td>2.71 (\pm) 0.05</td>
<td>2.86 (\pm) 0.01</td>
<td>2.68 (\pm) 0.05</td>
<td>2.95 (**)</td>
<td>2.69 (\pm) 0.16</td>
</tr>
<tr>
<td>12</td>
<td>2.68 (\pm) 0.16</td>
<td>2.97 (\pm) 0.15</td>
<td>2.97 (\pm) 0.02</td>
<td>2.64 (\pm) 0.11</td>
<td>2.87 (\pm) 0.12</td>
<td>2.60 (\pm) 0.16</td>
</tr>
</tbody>
</table>

*For these cases, hip impact velocities were not obtained because hip impact occurred at the same time or after the arm or hand impacted the mattress.

**Only one fall could be analyzed for this fall category, therefore, the standard deviation could not be determined.

Table 2. Mean values \([ \pm \text{S.D.} ]\) of trunk angle (deg.)

<table>
<thead>
<tr>
<th>Subject</th>
<th>S1-RX (\pm)</th>
<th>S1-AC (\pm)</th>
<th>S1-ARM (\pm)</th>
<th>W-RX (\pm)</th>
<th>W-AC (\pm)</th>
<th>W-ARM (\pm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>45</td>
<td>21.6 (\pm) 8.0</td>
<td>19.0 (\pm) 6.7</td>
<td>21.1 (\pm) 4.1</td>
<td>22.5 (\pm) 5.1</td>
<td>29.0 (\pm) 7.7</td>
<td>16.7 (\pm) 2.6</td>
</tr>
<tr>
<td>67</td>
<td>4.8 (\pm) 2.9</td>
<td>22.9 (\pm) 11.2</td>
<td>38.1 (\pm) 2.9*</td>
<td>2.6 (\pm) 2.0</td>
<td>13.7 (\pm) 2.3</td>
<td>35.2 (\pm) 7.7*</td>
</tr>
<tr>
<td>89</td>
<td>23.4 (\pm) 1.8</td>
<td>24.9 (\pm) 0.1</td>
<td>32.6 (\pm) 8.1</td>
<td>16.5 (\pm) 1.1</td>
<td>15.1 (\pm) 4.6</td>
<td>32.0 (\pm) 3.6</td>
</tr>
<tr>
<td>10</td>
<td>35.9 (\pm) 11.3</td>
<td>40.2 (\pm) 19.2</td>
<td>52.2 (\pm) 1.4*</td>
<td>18.9 (\pm) 12.9</td>
<td>34.3 (\pm) 7.2</td>
<td>54.1 (\pm) 9.8*</td>
</tr>
<tr>
<td>11</td>
<td>1.3 (\pm) 1.1</td>
<td>14.2 (\pm) 2.6</td>
<td>19.6 (\pm) 0</td>
<td>-6.0 (\pm) 3.8</td>
<td>1.9**</td>
<td>15.4 (\pm) 10</td>
</tr>
<tr>
<td>12</td>
<td>9.3 (\pm) 1.5</td>
<td>16.7 (\pm) 2.9</td>
<td>13.5 (\pm) 1.7</td>
<td>8.4 (\pm) 3.7</td>
<td>23.2 (\pm) 0.9</td>
<td>18.4 (\pm) 5.1</td>
</tr>
</tbody>
</table>

*For these cases, hip impact occurred at the same time or after the arm or hand impacted the mattress.

**Only one fall could be analyzed, therefore, the standard deviation could not be determined.

expressed by the trunk angle, defined as the angle between the trunk and the vertical. Since out-of-plane motion could not be detected, the trunk angle should be interpreted as the angle between the vertical and the projection of the trunk on the plane perpendicular to the camera.

Values for impact velocity at the mattress and at the floor level and trunk angle were averaged for each subject and for each fall category and used for statistical analyses. A fall was not analyzed if the subjects did not follow the instructions properly (for example if they stepped with the wrong foot on the force plate) or if some error in the recording process occurred. Also, for those falls in which the arm or hand impacted the mattress before the hip did, hip impact velocities were not obtained. The reason for this is as follows. When the hand or arm impacted first, this typically resulted in a discontinuity in the hip trajectory. Since this usually happened just before the hip impacted, it was not possible to obtain an accurate 'best fit curve' for this small portion of the position vs time curve, and therefore we could not obtain a good estimate of the hip impact velocity at the mattress level or at the floor level. We used a two-factor analysis of variance with repeated measures, with muscle-active vs muscle-relaxed and falling from standing vs from walking as trial factors, and hip impact velocity and trunk angle as dependent variables. We did not use gender as a grouping factor in this analysis. Instead, the effects of height and weight, which were higher for our male subjects than for the females, were tested separately by correlating the potential energy available in the fall with the average hip impact velocity for each subject. The potential energy available, \(mgh_{cm}\), was obtained by multiplying the subject's weight, \(mg\), by an estimate of the height of the subject's center of mass, \(h_{cm}\) (approximated by 0.51 times the subject's overall height; Baughman, 1983).

RESULTS

Hip impact velocity at the mattress ranged between 1.95 and 4.31 m s\(^{-1}\) (mean: 2.75 \(\pm\) 0.42 [S.D.], Table 1). Estimated values for floor impact velocity ranged between 2.14 and 4.79 m s\(^{-1}\).

![Fig. 2. Body configurations at impact for a muscle-active and a muscle-relaxed fall. The muscle-relaxed falls (bottom) resulted in a small reduction (7%) in hip impact velocity and a marked reduction (38%) of trunk angle (corresponding to a more vertical trunk) than the muscle-active falls (top). The insets show the EMG of the hamstrings; the arrows indicate the initiation of the fall.](image-url)
The configuration of the upper body at impact, expressed by the trunk angle, $\gamma$ (Figs 2 and 3, Table 2), ranged between $-8.62^\circ$ and $61.0^\circ$ (mean: $21.7 \pm 13.3$ [S.D.]). One of our six subjects impacted with the trunk slightly bent to the right (the falls were to the left), resulting in a negative trunk angle. No significant differences in trunk angle were found between falls initiated from walking and from standing.

Muscle activation during the falls significantly affected body configuration and moderately affected values for hip impact velocity (Fig. 2, Table 1). Muscle-relaxed falls resulted in a 38% reduction in the trunk angle at impact (corresponding to a more vertical trunk), compared to muscle active falls ($21.8 \pm 10.4^\circ$ vs $13.6^\circ \pm 11.2^\circ$; $p = 0.02$). Also, we found a 7% decrease in hip impact velocity at the mattress for relaxed vs muscle active falls ($2.86 \pm 0.41$ m s$^{-1}$ vs $2.66 \pm 0.36$ m s$^{-1}$; $p = 0.02$). Corresponding estimated values at the floor level were: $3.31 \pm 0.43$ m s$^{-1}$ vs $3.09 \pm 0.41$ m s$^{-1}$; $p = 0.04$.

Regarding the falls with an outstretched arm (fall types: ST-ARM, W-ARM), trunk angles were generally larger for these falls compared to falls with no outstretched arm or hand (Table 2). However, despite the instructions to 'break' the fall with the arm or hand, only two subjects were actually able to do so (Fig. 3). For these two subjects, mean values for trunk angle were larger compared to the remaining four subjects, for which hip impact occurred first, followed by contact with the arm or hand. Finally, the use of an outstretched arm did not have a clear effect on values for hip impact velocity (Table 1).

**DISCUSSION**

In this study, ranges of hip impact velocity and typical body configurations at impact associated with sideways falls from standing height were obtained. Average values for hip impact velocity at the floor and trunk angle were $3.17$ m s$^{-1}$ and $21.7^\circ$, respectively. Also, we investigated the effect on fall kinematics of protective reflexes, such as muscle activation and the use of an outstretched arm or hand to break the fall. Muscle-relaxed falls resulted in a small reduction in hip impact velocity and a marked decrease in trunk angle with respect to the vertical. Finally, we found that most of our subjects were not able, despite instructions, to break the falls with an outstretched arm.

To our knowledge, this is the first study to use human volunteers as subjects to study fall kinematics. The few studies to investigate falls with human subjects have focused primarily on fall initiation and balance recovery rather than on the descent phase of the fall (Chen et al., 1994; Do et al., 1982; Grabner et al., 1993; Luchies et al., 1994; Romick-Allen and Schulte, 1998).

Thus, we believe that the floor impact velocities reported here are more representative of an actual fall than our previous estimates using analytical models of falling based on energy conservation (Kroonenberg et al., 1995). In this study, we focused on sideways falls only, with impact on the hip or side of leg, since such a fall has the highest risk for hip fracture (Green-span et al., 1994; Hayes et al., 1993; Nevitt and Cummings, 1993).

Our data on hip impact velocity and body configuration at impact can be used to estimate the energy available at impact, which is an important parameter for the design of effective protective hip padding and force-attenuating flooring.

The surprising reduction in impact velocities found in muscle-relaxed falls appears to be the consequence of hip impact occurring closer to the floor in the muscle-relaxed case (Fig. 2). This finding suggests that falling relaxed reduces the injury potential of a fall. However, the smaller trunk angle at impact (corresponding to a more vertical trunk, Fig. 2) in muscle-relaxed falls could be expected to result in larger values for the effective moving mass of the body (Kroonenberg et al., 1995). This may reduce or even negate the lowering of hip impact force that would be an expected consequence of lower impact velocities in muscle-relaxed falls.

To our knowledge there is only one previous study that has investigated fall dynamics using human subjects. In their 'Pelvis Release' experiments, Robinovitch et al. (1991) simulated a fall to the side by releasing human subjects to fall from almost zero height on a force plate. The characteristics of the measured force records were used to estimate the stiffness of the tissue over the hip and the flexural stiffness of the total body. An important finding was that muscle-relaxed (trials resulted in a significant reduction of the stiffness, and therefore of the impact force, compared to muscle-active trials, thereby suggesting that falling relaxed reduces the injury potential of a fall. Our finding that falling relaxed results in a lower impact velocity supports the same conclusion.

The current study also provides insight into the energies associated with falling. Estimating the available energy just before impact by $\frac{1}{2}Mv^2$ with $M$ and $V$ the effective moving mass of the body and the vertical hip impact velocity, respectively, results in the following comparison. For a person of 66 kg (the average body mass for our six young subjects), and an effective mass corresponding to half the body weight (Robinovitch et al. 1991), the hip impact velocities we found result in 168 J of energy just before impact. Although these falls were not unintentional, the calculation indicates that a considerable amount of the total available energy has to be dissipated...
during the descent phase of a real fall from standing height, since an average of 589 J of potential energy was available initially. We considered protocols that would cause the subjects to fall unexpectedly and rejected them as too dangerous. The fall direction would have been uncertain for such falls, and therefore there would have been a chance of falling on the edge of the 0.6 m-high mattress and not on the floor. However, to minimize the effects of fall initiation on fall dynamics, we asked the subjects to fall as naturally as possible after they initiated the fall, as if the fall had occurred accidentally. It is also possible that human subjects fall differently when they are about to strike a mattress as opposed to a hard floor. We could not address this question, because we judged falls to the side on a hard floor to be unacceptably dangerous. Finally, these falls were performed by young individuals, since ethical and safety considerations ruled out the use of elderly subjects in these experiments.

A second limitation concerned the fact that the surface of the mattress was higher than the floor surface. We experimented with raising the walkway to the height of the mattress and decided against this configuration because subjects complained of the sensation of falling into a hole when they struck the mattress and kept their hands well below the walkway height. Instead, we left the mattress surface 12 cm higher than the force plate and extrapolated the hip trajectories to the force plate height in order to estimate the impact velocities that would have occurred had the subjects struck a floor level with the force plate. Evidence is given under Methods and Materials supporting the conclusion that the error in using such a model was not more than 2%. Third, because we were using only one video camera, motion that occurred out of the plane perpendicular to the video camera could not be detected. However, since we obtained the vertical component of the hip impact velocity only, out-of-plane motion of the hip marker did not affect these values. Out-of-plane motion of the hip marker due to rotation of the pelvis occurred just before impact in about 25% of the falls, resulting in disappearance of the hip marker from our view. We resolved this issue by making an estimation of the projection of the marker on the two-dimensional plane, using an outline of the subjects' left leg and trunk marked on a transparent sheet. Because of the hazardous nature of the experiments, we decided to keep the size of our subject population small and to limit it to young athletes, many of whom were active in sports in which falls are encountered (e.g. wrestling, gymnastics and football). These experiments were potentially dangerous in that sprains and even fractures can be an outcome of an actual fall, even for an intentional fall on a foam-rubber mattress. In fact, some subjects in our study complained of muscular soreness in the neck region (trapezius and sternocleidomastoid muscles) on the day following their participation in the experiments. The soreness disappeared within three days.

The narrow range of potential energies for our young subjects is the most likely reason that we found no relationship between the available potential energy and hip impact velocity. However, the differences we found between muscle-relaxed and muscle-active falls were statistically significant, and therefore we would expect to find the same result in a larger population of subjects.

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