Injury, Int. J. Care Injured (2004) 35, 417—424

INJURY
INTERNATIONAL JOURNAL OF THE CARE OF THE INJURED

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Influence of fall height and impact surface on biomechanics of feet-first free falls in children

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Accepted 4 February 2003

Summary *Objective*: The objectives of our study were to assess biomechanics associated with feet-first free falls in 3-year-old children and to investigate the influence of impact surface type and fall height on key biomechanical measures associated with injury risk. *Methods*: Repeatable feet-first free fall experiments were conducted in a laboratory mock-up environment using an instrumented Hybrid II 3-year-old test dummy. Impact surface type and fall height were varied to examine their influence on biomechanical measures. *Results*: Feet-first falls from short distances (27 in.) (0.69 m) were found to have a low risk of contact-type head injury, regardless of impact surface type. When comparing different types of impact surfaces in a 27 in. (0.69 m) fall, head acceleration associated with falls onto playground foam was significantly less than that associated with falls onto wood, linoleum or padded carpet. For falls onto playground foam, femoral compressive loads and bending moments were found to significantly increase as fall height increased. *Conclusions*: Impact surface type and fall height were found to influence biomechanics associated with injury risk in feet-first free falls as assessed through experimental mock-ups using an instrumented child test dummy. Feet-first falls from short distances (27 in.) (0.69 m) were associated with a low risk of contact-type head injury as assessed using HIC, irrespective of impact surface type.

Introduction

Falls are a common cause of childhood injury and are often falsely reported as a mechanism of injury in child abuse cases. The burden of determining the accuracy of a reported injury mechanism often rests on the treating physician. An understanding of fall biomechanics and influencing factors can serve as an objective aid when evaluating the compatibility between the stated cause and actual clinical findings; a key aspect when attempting to distinguish between paediatric abusive and non-abusive trauma.

Factors such as fall height and impact surface have been previously identified as factors

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doi:10.1016/S0020-1383(03)00062-7

influencing injury potential and severity.[4,19] Studies focusing on playground-specific falls have also found fall height and impact surface characteristics to have an important role in injury incidence and severity.[2,16,20,21] *A systematic means of studying each of these factors in isolation could be beneficial in attempting to determine their contributory nature towards injury risk*. Experiments designed to study a specific fall mechanism in a human analogue or anthropomorphic test dummy (ATD) is one way to achieve this goal.

Falls have been confirmed as a primary cause of injury in children, resulting in the greatest number of emergency department visits and hospital admissions[8] as compared to other injury causes. Falls have also been found to be responsible for generating the greatest portion of head injuries and fractures in children.[1,5,13,25,28] According to Smith et al., falls were the most common cause of traumatic death in children under the age of 15.[26] In a 1986 study of children who were hospitalized due to brain injury, it was estimated that nearly 30% of childhood injury-related deaths resulted from head injury, half of which were associated with falls.[12] Duhaime's study of head injury found household falls to be a common mechanism and that fall heights influenced injury types.[5]

Foust et al. investigated impact tolerances of the head and lower extremities in free-fall accidents occurring in children, female adults and the elderly populations.[7] These researchers relied upon scene investigations, theoretical biodynamic calculations, experimental free falls using ATDs and 2D computer simulation models to predict key biomechanical impact measures which have previously been correlated with injury risk. Researchers showed a relation between surface types and severity of injury. They also found that head-first falls from greater than 10 feet (3.05 m) onto a rigid surface were predicted to result in skull fracture or concussion, and at least AIS = 2 injuries for adults and children.

Since lower extremity and head injuries have been found to be common in free falls, our study sought to estimate femoral loading, head acceleration and head injury criteria (HIC) associated with free falls in children with a vertical initial posture and feet-first landing. Our study investigated the influence of common impact surfaces and fall height on key biomechanical measures and injury risk in children.

Methods

Repeated fall experiments were conducted using a paediatric test dummy to investigate feet-first free falls in children. A Hybrid II 3-year-old ATD was used to represent the fall victim. The head and pelvis were instrumented with three uni-axial accelerometers arranged with one accelerometer for each axis. The pelvis was modified to accommodate the mounting of three uni-axial accelerometers. The upper legs of the ATD used a metal rod to simulate the femur and was instrumented with four strain gages to measure femoral loading. Three of the gages were positioned to measure axial and bending load, and the fourth gage measured torsion load. Overall gross motion kinematics were also captured using videography.

Data acquisition and analysis

A LabVIEW program was created for the purpose of data acquisition. The data were collected at a sampling rate of 1000 Hz.

Data were analyzed to obtain the peak values of each measure, duration of loading spike, and impulse. The head acceleration data were used to calculate HIC which accounts for duration of acceleration exposure. The formula for the HIC is defined as:

$$\mathrm{HIC} = (t_2 - t_1)\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\,\mathrm{d}t\right]^{2.5}$$

where $a(t)$ is the acceleration and t_1 and t_2 are the start and finish times of the acceleration spike. In our study the time between t_1 and t_2 was taken as 36 ms so that our measures were comparable to established injury criteria. HIC is commonly used to assess risk of head injury in motor vehicle crashes,[22] as well as to evaluate the safety of materials such as playground surfaces, and motorcycle and sports helmets.[14,24]

Procedures

The dummy was dropped to emulate a feet-first free fall on various impact surfaces (wood, padded carpet, linoleum and playground foam) from the same initial height and position. Additionally, drops from three different heights (27, 47 and 64 in. (0.69 m, 1.19 m, 1.63 m) as measured from the center of mass of the ATD) were performed on the playground foam to measure the effect of initial fall height.

As recommended by the ATD manufacturer, joints were calibrated so that the friction was sufficient to just support the weight of the limb (1 g). The ATD was suspended by an eyebolt in the top of the head to a release mechanism attached to the ceiling. This allowed repeatability for the height of the fall, as well as the release. The ATD was placed in a standing posture with arms extended above the head. The orientation of each joint was measured using a goniometer and the ATD was reset to the same position at the beginning of each drop for maximum repeatability. Each testing configuration was repeated three times.

The ATD impact surfaces used in our study represented actual indoor and outdoor surfaces; the tested surfaces included padded carpet, wood, linoleum and playground foam. Impact surfaces other than playground foam were adhered to a 6 ft × 6 ft (1.83 m × 1.83 m) platform constructed of 3/4 in. (19 mm) plywood flooring covering 2 in. × 4 in. (5.1 cm × 10.2 cm) joists positioned 16 in. (0.41 m) on centre. The playground foam surface consisted of a series of 2 ft × 2 ft (0.61 m × 0.61 m) tiles (2 in. thick) (5.1 cm) placed over a concrete sub-floor and arranged to create a 6 ft × 6 ft (1.83 m × 1.83 m) bordered surface.

Impact surface properties

The coefficients of friction were measured between each impact surface, the ATD and the ATD shoe. Both the static and dynamic coefficients of friction were experimentally determined (Table 1).

The coefficient of restitution (COR), a measure of how efficiently energy is returned to an object after impact, was also experimentally measured for each impact surface (Table 2). A COR of 1.0 is perfectly elastic, while a COR of 0 is inelastic having no rebound on impact.

Table 1 Coefficient of friction properties of impact surfaces

Impact surface	ATD (static/dynamic)	Shoe (static/dynamic)
Playground foam	1.23/0.97	0.71/0.70
Carpet	1.23/1.14	0.58/0.51
Linoleum	1.25/1.00	0.43/0.43
Wood	1.04/0.82	0.42/0.29

Table 2 Coefficient of restitution properties of impact surfaces

Impact surface	COR
Playground foam	0.57
Carpet	0.33
Linoleum	0.12
Wood	0.12

Statistical analysis

Data from each of the conditions were considered as replicates in order to measure variability within each. For the statistical comparisons of surface conditions, a Kruskal—Wallis test was used for each of the outcomes as the necessary statistical assumptions of normality and homoscedacity were violated. For the statistical comparisons of height conditions, an analysis of variance was used for each outcome since all necessary statistical assumptions were met. Following statistically significant omnibus tests for either the Kruskal—Wallis or analysis of variance, pairwise post-hoc comparisons were done using an overall controlled alpha. Statistical significance was defined as $P \leq 0.05$.

Results

Height of fall (playground foam surface)

The influence of fall height on biomechanical measures was evaluated by conducting falls from three different heights (27, 47 and 64 in.) (0.69 m, 1.19 m, 1.63 m) onto a playground foam surface. The height of the fall was measured from the impact surface to the center of mass of the ATD.

Fall dynamics varied across the fall height scenarios. For the 27 in. (0.69 m) fall (Fig. 1), the ATD impacted feet-first, the knees then flexed and the torso rotated rearward impacting the surface. For the 64 in. (1.63 m) fall, the feet impacted the surface, the knees flexed and ankles dorsiflexed allowing the knees to contact the impact surface while the pelvis continued moving downward. The ATD then rebounded upward and rearward, extending at the hip and falling backward onto the impact surface. A similar fall dynamic was also observed in the 47 in. (1.19 m) free fall.

Figure 1 Vertical feet-first free fall dynamics of 3-year-old ATD from 27 in. (0.69 m) height onto playground foam surface.

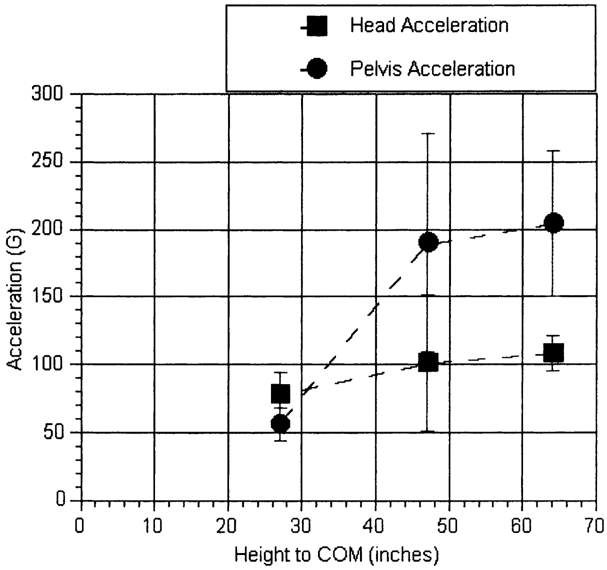

Figure 2 Head and pelvis acceleration vs. fall height for playground foam impact surface.

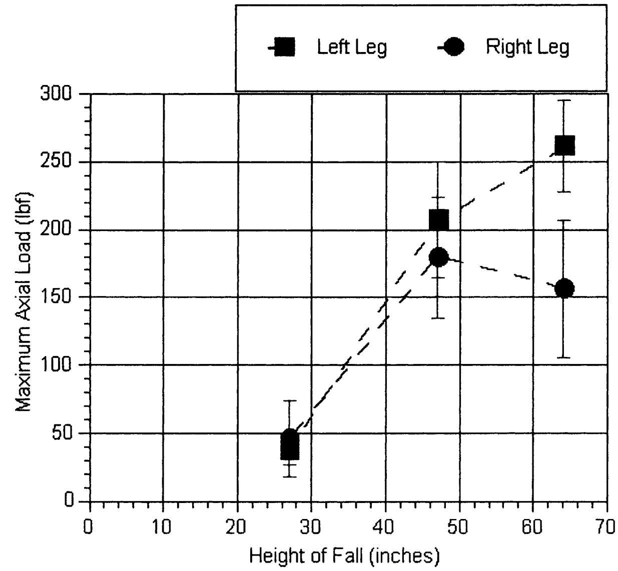

Figure 4 Left and right femur peak axial loading vs. fall height for playground foam impact surface.

There was a general increasing trend in the head and pelvis acceleration as the fall height increased (Fig. 2). The mean peak head acceleration increased by 38% as the fall height was increased from 27 to 64 in. (0.69 m to 1.63 m). However, a statistically significant difference in head acceleration was not found across varying fall heights. Pelvic acceleration generally increased with fall height and a significant ($P = 0.033$) difference was found when comparing the 27 (0.69 m) and 64 in. (1.63 m) fall height scenarios.

HIC values did not show a specific trend relative to fall height (Fig. 3). HIC values associated with the 64 in. (1.63 m) fall had large variability (mean = 76, S.D. = 63).

An increase in peak femoral compressive loading was observed with increasing fall height (Fig. 4). Both the left ($P \leq 0.001$) and right ($P = 0.018$) legs incurred a statistically significant difference in compressive loading across varying fall heights. Femoral bending moments were also found to increase with increasing fall height (Fig. 5). There was a significant difference ($P = 0.022$) in bending moment measured in the right leg when comparing varying fall height scenarios. Femur torsional moments (Fig. 6) were not found to vary significantly across fall heights. In summary, our femoral loading data suggests that compressive (axial) and bending loading increases with height when falling feet-first onto a playground foam surface.

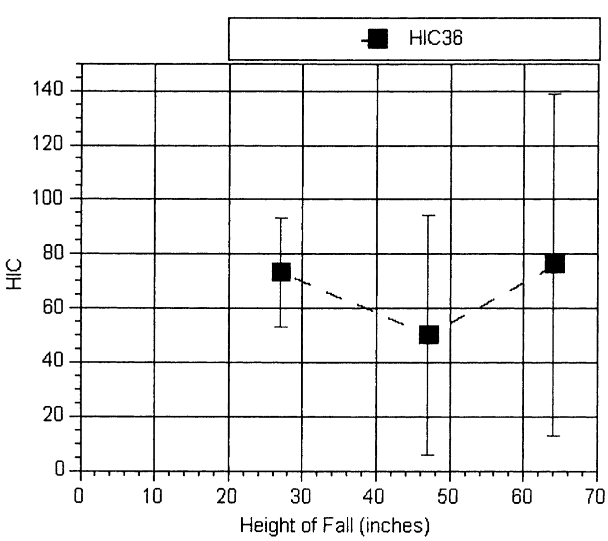

Figure 3 HIC vs. fall height for playground foam impact surface.

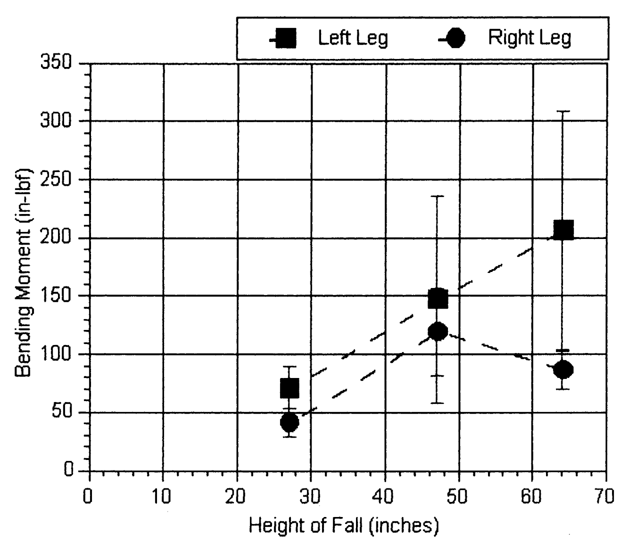

Figure 5 Left and right femur peak bending moment vs. fall height for playground foam impact surface.

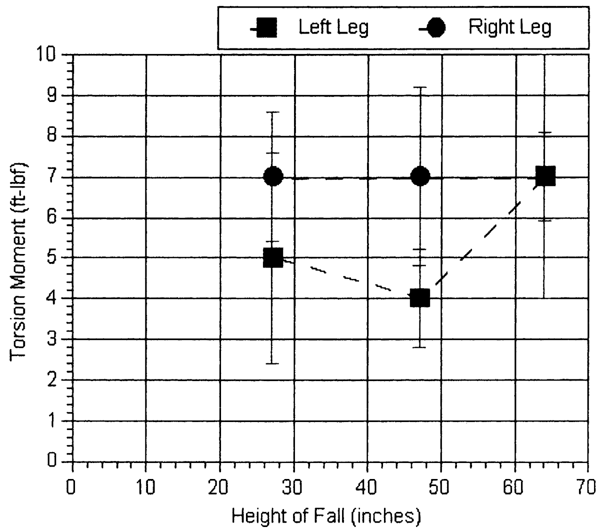

Figure 6 Left and right femur peak torsional moment vs. fall height for playground foam impact surface.

Impact surface

The influence of impact surface on biomechanical measures was evaluated by conducting falls from a fixed height of 27 in. (0.69 m) (measured from the ATD center of mass) onto various impact surfaces.

A significant difference was found in head acceleration (Fig. 7) generated in falls onto playground foam as compared to all other tested impact surfaces ($P = 0.05$). Falls onto playground foam were found to generate a lower head acceleration than other surfaces. When comparing HIC values (Fig. 7)

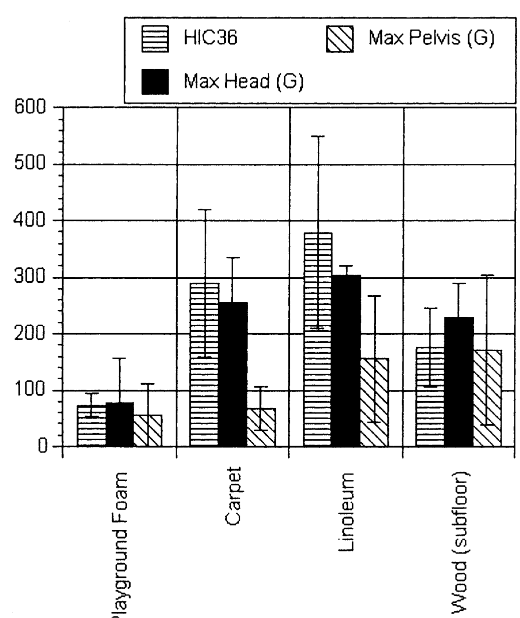

Figure 7 HIC, peak head acceleration and peak pelvis acceleration for 27 in. fall height onto playground foam, carpet, linoleum and wood impact surfaces.

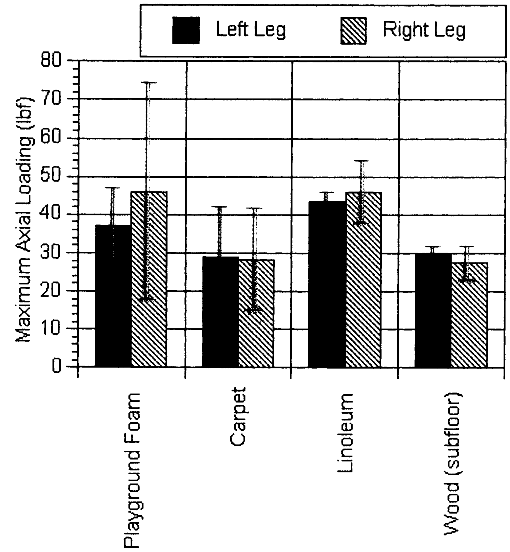

Figure 8 Peak femur axial loading (left and right leg) for 27 in. (0.69 m) fall height onto playground foam, carpet, linoleum and wood impact surfaces.

between linoleum and playground foam, a significant difference was found ($P = 0.05$). The highest HIC values were associated with the linoleum impact surface. No statistically significant difference was found in pelvis acceleration across the various impact surfaces.

Femoral compressive loading (Fig. 8) was not found to vary significantly across the various impact surfaces. However, left leg femoral bending loads (Fig. 9) associated with falls on playground foam as compared to other tested surfaces (carpet, linoleum and wood) were found to be significantly

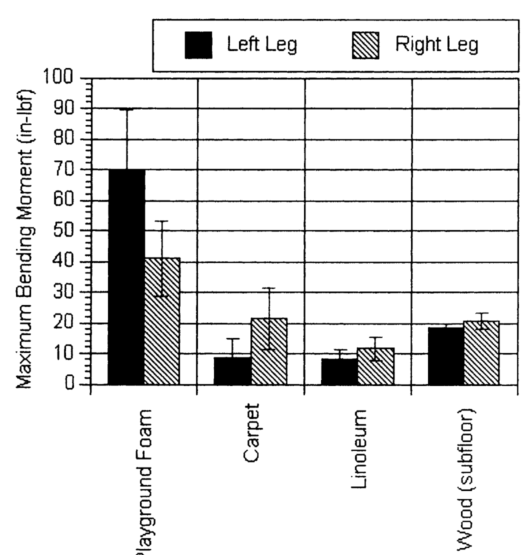

Figure 9 Peak femur bending moment (left and right leg) for 27 in. (0.69 m) fall height onto playground foam, carpet, linoleum and wood impact surfaces.

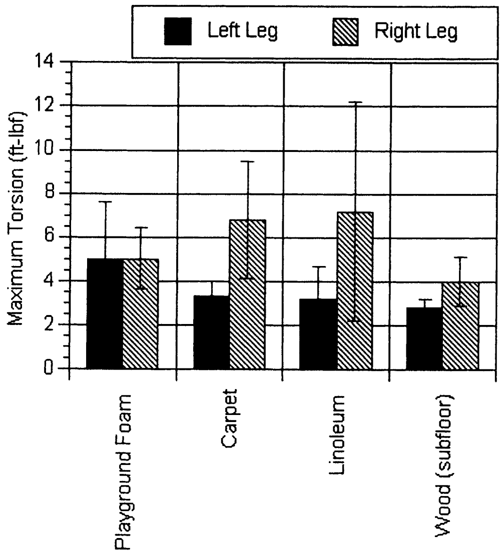

Figure 10 Peak femur torsional moment (left and right leg) for 27 in. (0.69 m) fall height onto playground foam, carpet, linoleum and wood impact surfaces.

higher ($P = 0.025$). Similar significant differences were also found in right leg bending (Fig. 9) across linoleum and playground foam impact surfaces ($P = 0.045$). Significant differences were not observed in torsional loading across the various impact surfaces Fig. 10.

Discussion

As found in previous studies, our findings indicate that fall height influenced free fall biomechanics and therefore would likely affect injury risk.[16,28] In our experiments, pelvis acceleration and femoral compressive loading significantly increased with fall height for feet-first falls onto playground foam. However, HIC values did not significantly change with increasing fall height when impacting feet-first onto playground foam. This finding is also reflective of the large variation that was associated with the higher (64 in.) (1.63 m) fall height. Such variation in HIC values are likely due to variations in impact kinematics and landing positions associated with falls from higher heights. Additional experimental tests should be conducted to further investigate these variations in landing position and HIC from higher distance falls and onto surfaces other than playground foam.

For a short distance, 27 in. (0.69 m) fall height, surface type was found to influence biomechanical measures and will likely affect injury risk. Head acceleration, HIC and lower extremity bending loads were significantly different when comparing falls onto linoleum versus falls onto playground foam. Playground foam proved to be the best surface in terms of reducing contact-type head injury risk, while linoleum proved to be the worst. These findings highlight the importance of impact surface in predicting likelihood of head injury in a short distance fall (Fig. 10). These experimental findings are supported by similar epidemiological fall studies which also found impact surface to influence injury outcome.[2,16,20]

Although playground foam had a beneficial effect when considering contact head injury risk in 27 in. (0.69 m) short distance falls, the same benefits were not realized when considering lower extremity injury risk. In fact, as compared to the other surfaces, playground foam offered no reduction in lower extremity loading at the 27 in. (0.69 m) fall height, and actually was associated with increased bending. The increased bending moment associated with the playground foam was likely due to the foot impacting and ''sticking'' (as opposed to sliding free), creating a greater resistive moment about the lower extremity. Such a response is greatly dependent upon surface friction. (Playground foam has a higher coefficient of friction than other tested surfaces.)

While it is of interest to compare injury risk and trends in biomechanical measures associated with various impact surfaces and heights, it is not possible to definitively indicate the clinical outcome associated with such loading or accelerations. Biomechanical measures must be compared to injury tolerance levels for the studied population in order to predict injury risk, but unfortunately relatively little data exists related to injury tolerance in 3-year-old children.

Recently however, NHTSA attempted to adapt the HIC established for mid-sized males to various sized crash victims.[11] To do so, scaling factors must account for both geometric and material property differences. Material properties specific to the cranial sutures were chosen as the key scaling property in children, reflecting resistance to skull fracture in children. Accounting for these factors, NHTSA proposed a HIC value of 900 for 3-year-old children. The probability of skull fracture (AIS ≥ 2) associated with the proposed NHTSA HIC limit is estimated to be 47% across all test dummies.

When comparing to proposed NHTSA HIC thresholds for 3-year-old children (HIC = 900), none of the falls from a 27 in. (0.69 m) height onto various impact surfaces exceeded the HIC threshold, and would therefore be associated with a low probability of contact-type head injury. These findings are similar to the findings of previously conducted epidemiological studies indicating that severe head injury does not typically result from short distance falls.[9,15,23] Similarly, feet-first falls from 47 (1.19 m) and 64 in. (1.63 m) onto playground foam generated HIC values that were less than the proposed NHTSA HIC threshold of 900 for 3-year-old children, and would therefore also be associated with a reduced risk of contact-type head injury. It is important to note that HIC is representative of contact-type head injuries (e.g. skull fractures) and does not reflect head injuries associated with angular acceleration.

Unfortunately injury tolerance data for the femur in 3-year-old children is scarce. In quasistatic testing of a single 2.5-year-old child, Martin and Atkinson found a cadaveric femur threshold bending moment of 465 in. lb (2.66 N-m).[17] None of the bending moments generated in our experimental falls exceeded this threshold established by Martin and Atkinson. Stanley evaluated 17 cadaveric femurs from children 6 years and younger, and found a compressive strength range of 44—205 lb/in.^{2} (303—1413 kPa).[27] In this case, strength reflects the point at which bone yielding and permanent deformation would begin. Using data for the 3-year-old, Stanley proposed a compressive fracture limit of 198 lb (881 kPa). In comparison with our results, falls from 47 (1.19 m) and 64 in. (1.63 m) onto playground foam generated loads that exceeded the compressive fracture limit reported by Stanley, and therefore could be associated with the potential for fracture. None of the falls from 27 in. (0.69 m) onto various test surfaces exceeded this femur compressive load limit. However, it is important to note that loads measured in our study were not cadaver based and were instead measured using a human analog. Correlation studies between the Hybrid II ATD femur and cadaveric results would be necessary in order to make such a direct comparison to validate the biofidelity of the ATD. It is therefore important to note that fracture prediction provides only limited value in this case. Information regarding trends can, however, guide the clinician in evaluating fracture type and severity in relation to distance fallen, surface type and landing position.

This study provides preliminary biomechanical data associated with feet-first free fall events in 3-year-old children. It is important to note however that these measures are derived from a Hybrid II ATD which only approximates a 3-year-old child. Since biomechanical response data on children are scarce, child ATDs are typically less biofidelic than their adult counterparts. Scaling techniques from adult to child, based upon geometry and mass, are often used in the development of smaller ATDs.[18] Additionally, despite a few studies attempting to extend the use of Hybrid II test dummy to low energy events,[3,6,10] the development of the Hybrid II ATD was intended for high energy impact events such as motor vehicle crashes. The application of this ATD to the study of fall biomechanics should therefore be limited to investigating trends in biomechanical measures. Moreover, lower extremity response to loading, assessed through the addition of strain gages to the structure representing the ATD femur, has not been previously validated for biofidelity. Therefore, lower extremity loads described in this study should be used primarily for assessing trends in femoral loading across various fall conditions.

Future studies should include a sensitivity analysis on initial, pre-fall, position. In other words, how does the child's initial position affect fall biomechanics and injury risk. Also, future studies should attempt to conduct a greater number of fall experiments for each test condition. Improvements in the biofidelity of paediatric ATDs would also greatly enhance the ability to predict injury risk in common childhood falls. Injury risk associated with head injuries resulting from angular accelerations should also be investigated in future fall studies.

Conclusions

Our findings showed that fall height, impact surface and landing position can influence injury risk in feet-first free falls. Despite limitations in the extrapolation from test dummy measures to human tolerance, relationships between fall environment parameters and key injury biomechanical measures were clearly observed. Our findings indicate that there is a low risk of contact-type head injury in feet-first falls from a short distance (27 in. as measured to the center of mass of ATD) (0.69 m), regardless of impact surface type. Despite the low risk of contact-type head injury in short distance falls, impact surface type was found to play a significant role in contact-type head injury risk, even in a feet-first fall orientation. Height of fall was found to significantly affect the risk of leg injury in falls onto playground foam. These findings further highlight the affect that subtle fall environment differences can have on injury risk, and point to the need for detailed histories when attempting to differentiate between abusive and non-abusive injuries.

Acknowledgements

Funding for this work was provided by: The Whitaker Foundation; CDC—Center for Injury Research and Control, and Children's Hospital of Pittsburgh.

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