Assessment of Head Injury Risk Associated With Feet-First Free Falls in 12-Month-Old Children Using an Anthropomorphic Test Device

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Background: Short distance falls are a common false history provided in cases of child abuse. Falls are also a common occurrence in ambulating young children. The purpose of this study was to determine the risk of head injury in short distance feet-first free falls for a 12-month-old child.

Methods: Feet-first free falls were simulated using an anthropomorphic test device. Three fall heights and five surfaces were tested to determine whether changing fall environment characteristics leads to differences in head injury risk outcomes. Linear head accelerations were measured and angular head accelerations in the anterior-posterior direction were determined. Head injury criteria values and impact durations were also determined for each fall.

Results: The mean peak linear head acceleration across all trials was 52.2*g*. HIC_{15} values were all below the injury assessment reference value. The mean peak angular head acceleration across all trials was 4,246 rad/s^2. Impact durations ranged from 12.1 milliseconds to 27.8 milliseconds. In general, head accelerations were greater and impact durations were lower for surfaces with lower coefficients of restitution (a measure of resiliency). In falls onto wood and linoleum over concrete, the ground-based fall was associated with greater accelerations than the two higher fall heights.

Conclusions: Results show that fall dynamics play an important role in head injury outcome measures. Different fall heights and impact surfaces led to differences in head injury risk, but the risk of severe head injury across all tested scenarios was low for a 12-month-old child in feet-first free falls.

Key Words: Falls, Child abuse, Head injury, Biomechanics.

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Child abuse is the leading cause of trauma-related fatalities in children less than 4 years of age.[1] In the US in 2005, there were approximately 899,000 victims of child abuse. There were also approximately 1,460 fatalities because of child abuse with 76.6% of these cases involving children aged 3 years or less.[2] These numbers may be underestimated because it has been suggested that as many as 50% to 60% of deaths related to child abuse go unrecorded.[3]

Inflicted traumatic brain injury (TBI) is the leading cause of child abuse fatalities.[4–6] In children less than 1 year of age, serious TBI is more likely to be the result of inflicted than noninflicted injury.[5] In children aged 0 to 4, falls are the leading cause of TBI.[7] However, falls are a common false history given by caretakers to cover up inflicted trauma.[8,9] In children aged 0 to 4, approximately 32% of TBI deaths are caused by inflicted trauma, whereas only 3% of TBI deaths are caused by falls.[7] Because short distance falls are a common occurrence in young ambulating children, a clearer understanding of the injury potential associated with common fall scenarios could prove useful. Clinicians are commonly asked to determine whether a child's injuries are consistent with the given cause of the injuries, and the diagnosis of abuse may hinge on this decision. Currently, there is little scientific evidence in the field of injury biomechanics to aid in this decision making. Specific information regarding injury risk associated with common falls falsely reported in child abuse can aid clinicians in distinguishing between inflicted and noninflicted injuries. Early detection of abuse may lead to prevention of further escalating injuries and, in some cases, prevent the death of the child.[6]

As a first step in understanding the kinematics of simple falls and how they relate to head injury risk, feet-first free falls were experimentally simulated using an instrumented anthropomorphic test device (ATD), or human surrogate, representing a 12-month-old child. Although feet-first falls do not maximize head injury risk, anecdotal evidence indicates that feet-first falls are common in young children and have also been given as an explanation of injury causation in child abuse legal cases. The purpose of this study was to investigate the risk of head injury for a 12-month-old child in feet-first free falls and to determine the effect of varying fall height and impact surface on head injury risk.

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METHODS

Test Setup

A Child Restraint Air Bag Interaction (CRABI) 12-month-old ATD (First Technology Safety Systems, Plymouth, MI) was suspended in a vertical initial posture and dropped to simulate a free fall. The ATD was suspended from a rope secured to the neck and then dropped from a releasing mechanism with an external trigger to assure repeatability of the fall (Fig. 1). The ATD represents a 50th percentile 12-month-old child in terms of overall height and mass, as well as geometric and inertial properties of individual body segments. The ATD was instrumented with four accelerometers (Endevco, Model 7264-2000T2) positioned within the head. Three accelerometers were placed triaxially at the center of mass of the head and a fourth was located 1.6 inches (4.1 cm) posterior to the head center of mass. In addition to measurement of linear accelerations at the center of mass of the head using the accelerometers placed there, the fourth accelerometer offset from the center of mass allowed for calculation of angular head acceleration.

The ATD was dropped from heights of 18 inches, 27 inches, and 47 inches (45.7 cm, 68.6 cm, and 119.4 cm) measured from the ground to the center of mass of the ATD, which is located 18 inches vertically above the plantar surface of the feet. Therefore, the 18-inch fall represents a ground-based fall. The 27-inch and 47-inch fall heights can be thought of as representative of falls from a short stool and a chair, respectively.

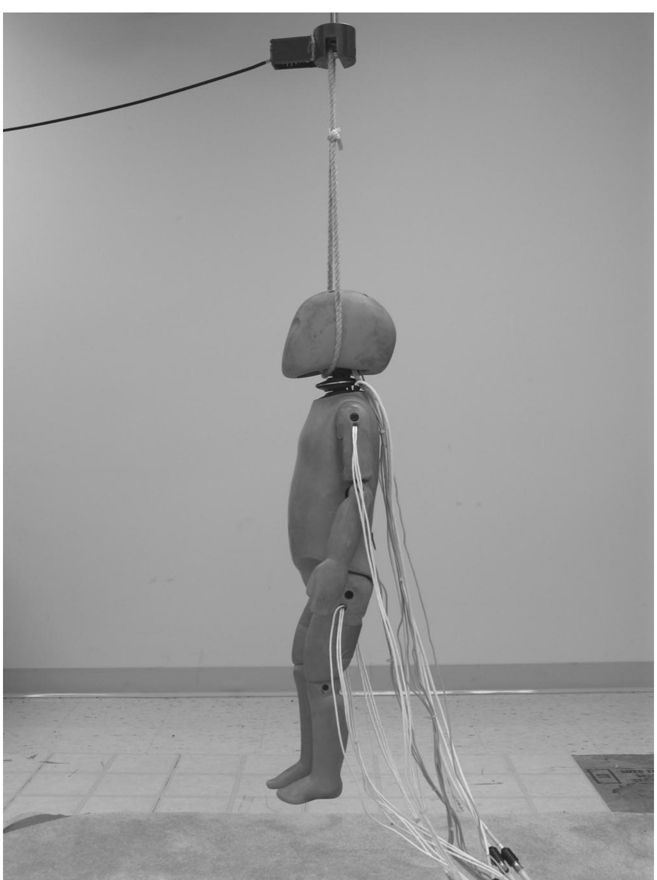

Fig. 1. ATD *suspended for fall experiments.*

Before each fall, ATD joint angles were adjusted using a goniometer to ensure repeated positioning for all testing. Joints were calibrated to manufacturer specifications whereby the joint was tightened until the friction was just sufficient to support the weight of the limb.

Five different impact surfaces were tested: linoleum over wood, playground foam, padded carpet, wood, and a linoleum-tiled concrete floor. All surfaces except the linoleum-tiled concrete floor were placed over a 6 feet × 6 feet (183 cm × 183 cm) wooden platform. The platform consisted of 3/4 inch (1.9 cm) plywood covering 2 inch × 4 inch (5.1 cm × 10.2 cm) joists spaced 16 inches (40.6 cm) apart. The linoleum over wood was no-wax self-adhesive vinyl flooring (0.039 inch or 1-mm thick) adhered to the platform. The linoleum tile over the concrete floor (different from the linoleum used over the wood floor) was 1/8-inch (0.3175 cm) thick. The playground foam surface consisted of 2 feet × 2 feet (61.0 cm × 61.0 cm) tiles, 2-inch (5.1 cm) thick. The carpet was open loop and 1/2-inch (1.3 cm) thick with 3/8-inch (1.0 cm) thick foam padding underneath. A layer of 3/4-inch (1.9 cm) thick plywood served as the wood surface. To further describe each impact surface, the coefficients of friction (COF) and coefficients of restitution (COR) (a measure of resiliency or elasticity) were measured (Table 1). The measured COF values describe the interaction between the impact surface and a barefoot ATD. It should be noted that COF values depend on the surfaces in contact and would likely be different if the ATD were wearing shoes or socks. However, fall experiments were performed without any ATD footwear and thus the COF was determined based upon the interaction between the ATD (without shoes or socks) and the impact surface.

Data Acquisition

A LabView program was created for data acquisition. Accelerometer data were sampled at 10,000 Hz and filtered at 1,000 Hz according to SAE J211 standards.[10] The filter was

Table 1 Coefficients of Friction and Restitution for Each Impact Surface

Impact Surface	Static Coefficient of Friction†	Coefficient of Restitution†
Padded carpet	1.10	0.58
Playground foam	0.88	0.47
Linoleum over wood*	0.87	0.40
Linoleum over concrete*	0.78	0.41
Wood	0.70	0.45

* Linoleum over wood is a different product from linoleum over concrete. Linoleum over wood has a rougher surface.

† A lower coefficient of friction indicates a more slippery surface. A higher coefficient of restitution indicates a more elastic surface.

a 4th order low-pass Butterworth filter. Falls were videotaped at a 30 Hz frame rate to capture fall dynamics. Ten drops were performed for each test scenario based upon a previous power analysis.

Data Analysis

Linear head acceleration was evaluated by examining both the maximum resultant acceleration at the head center of mass and Head Injury Criteria (HIC)[11] values which were calculated using

$$HIC = (t_2 - t_1)\left[\frac{1}{(t_2 - t_1)}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}_{max}$$

where $a(t)$ is the resultant linear head acceleration, and t_1 and t_2 are the start and finish times of the acceleration spike. HIC values were calculated over 15 milliseconds durations (HIC_{15}) and compared with proposed injury criteria.[12]

Anterior-posterior (AP) angular head accelerations were determined according to the following equation:

$$a = \frac{a_1(t) - a_2(t)}{d}$$

where $a_1(t)$ is the linear head acceleration in the superior direction measured at the center of mass of the head, $a_2(t)$ is the linear head acceleration in the superior direction measured by the accelerometer posterior to the center of mass of the head, and d is the horizontal distance between the two accelerometers. Angular accelerations were compared with published head injury thresholds. Impact durations were also determined for each fall. Impact durations were determined from the AP linear acceleration trace as the time duration of the acceleration peak with the greatest magnitude (where head impact occurred).

Each of the outcome variables was analyzed separately using one-way analysis of variance tests to determine whether surface type or fall height led to significant differences in the outcome measures. One way analysis of variances were used, because the interaction effect of height and surface was not of interest. Rather, the purpose of this study was to understand how each variable independently affects the fall kinematics and head injury outcome measures. *Post hoc* Tukey tests were conducted to further examine where significant differences occurred. Statistical significance was set at $p \leq 0.05$.

RESULTS

Fall Dynamics—Qualitative Assessment

Differences Due to Changing Fall Height

In the 18-inch falls, after release, the ATD fell to a squatting position with hips and knees flexed, and then fell rearward, rotating about the feet, first contacting the surface with the posterior pelvis and then the posterior aspect of the head (Fig. 2). The ATD kinematics upon impact were similar for the 27-inch and 47-inch falls. In these falls, the ATD contacted the surface feet-first, which led to hip and knee flexion, followed by rebounding of the entire ATD body

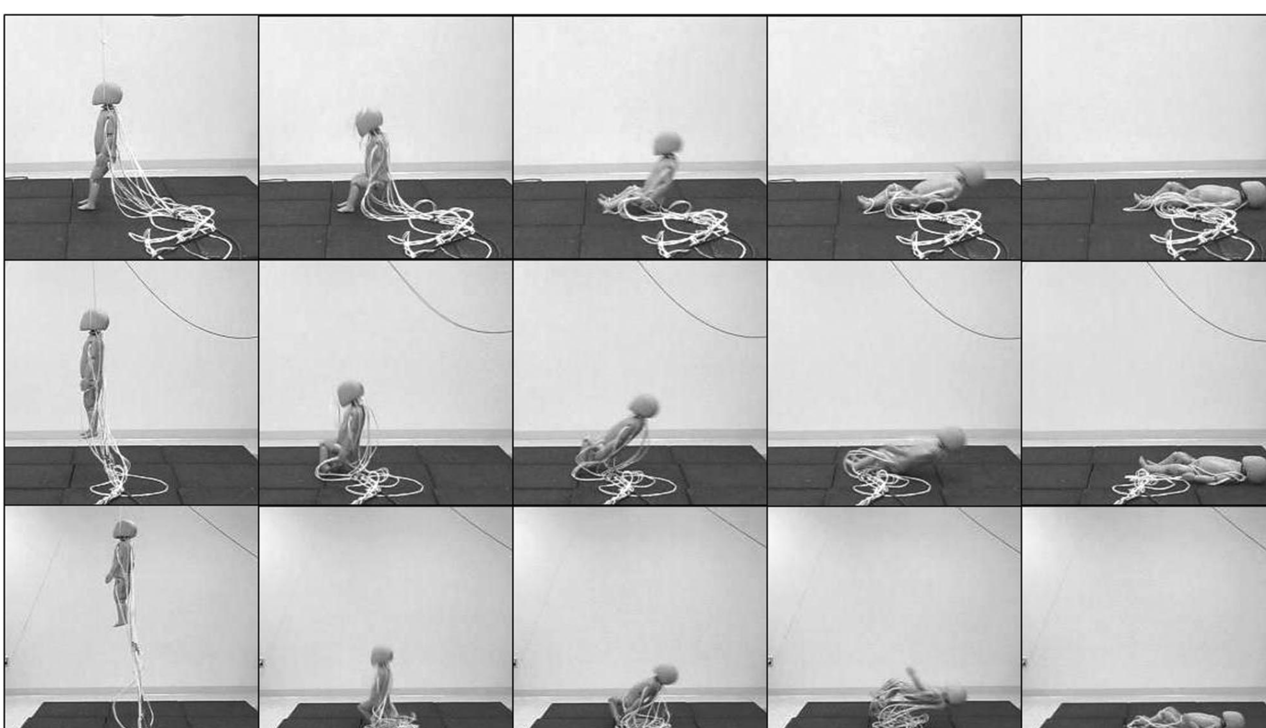

Fig. 2. *Representative 18, 27, and 47 inches (distance from ground to* ATD *center of mass) falls onto playground foam.*

upward and rearward off of the impact surface. This led to posterior head and torso impact with the ground surface almost simultaneously. In lower height falls, the ATD most often fell directly rearward, but with increasing fall height, the dynamics became less predictable, with the ATD falling to its side and on its back.

Differences Due to Changing Impact Surface

For the two higher fall heights (27 inches and 47 inches) onto linoleum over wood, carpet, and playground foam, the ATD's feet tended to "stick" to the surface upon impact, whereas in falls onto wood and linoleum-tiled concrete, the ATD's feet tended to slide forward after initial impact with the ground surface.

Linear Head Acceleration

The mean peak resultant linear head acceleration across all trials was 52.2*g* (95% confidence interval [CI], 49.0–55.5) (Fig. 3). The 18-inch falls onto linoleum over concrete produced the largest values, with a maximum of 130.6*g*. Significant differences in peak resultant linear head acceleration because of fall height occurred only for the wood and linoleum over concrete surfaces (Table 2). For these two surfaces, the 18-inch fall height was associated with significantly greater linear head accelerations than either the 27-inch or 47-inch heights. There were significant differences in peak resultant linear head acceleration for different impact surfaces across all fall heights tested (Table 3).

Head Injury Criteria

The mean HIC_{15} value across all trials was 67 (95% CI, 62–73) (Fig. 4). The maximum HIC_{15} was 173 and occurred during an 18-inch fall onto linoleum over concrete. This value is well below the injury assessment reference value of 390 for the 12-month-old ATD.[12] There were no significant differences in HIC_{15} values because of fall height for the linoleum over wood and playground foam surfaces. For falls onto carpet, wood, and linoleum over concrete surfaces, the

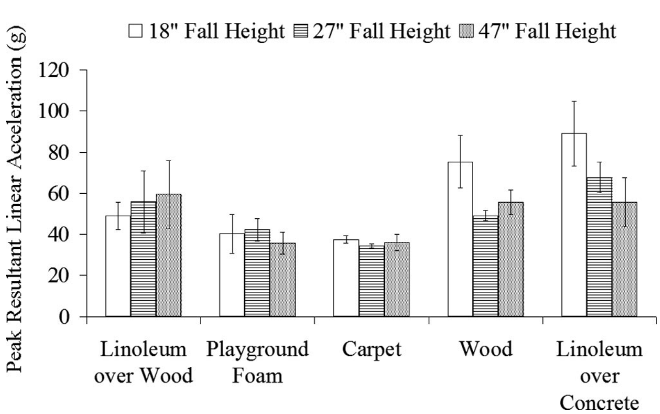

Fig. 3. *Peak resultant linear head accelerations for various impact surfaces and fall heights (measured from ground to* ATD *center of mass).* Error bars *represent 95%* CI.

Table 2 Significant Differences in Peak Resultant Linear Head Accelerations for Various Heights and a Fixed Impact Surface

	18 inches	27 inches	47 inches
Linoleum over wood			
18 inches			
27 inches			
47 inches			
Playground foam			
18 inches			
27 inches			
47 inches			
Carpet			
18 inches			
27 inches			
47 inches			
Wood			
18 inches		X	X
27 inches	X		
47 inches	X		
Linoleum over concrete			
18 inches		X	X
27 inches	X		
47 inches	X		

"X" indicates significant difference ($p < 0.05$).

Table 3 Significant Differences in Peak Resultant Linear Head Accelerations for Various Surfaces and a Fixed Fall Height

	Linoleum Over Wood	Playground Foam	Carpet	Wood	Linoleum Over Concrete
18 inches					
Linoleum over wood				X	X
Playground foam				X	X
Carpet				X	X
Wood	X	X	X		
Linoleum over concrete	X	X	X		
27 inches					
Linoleum over wood			X		
Playground foam					X
Carpet	X			X	X
Wood			X		X
Linoleum over concrete		X	X	X	
47 inches					
Linoleum over wood		X	X		
Playground foam	X			X	X
Carpet	X			X	X
Wood		X	X		
Linoleum over concrete		X	X		

"X" indicates significant difference ($p < 0.05$).

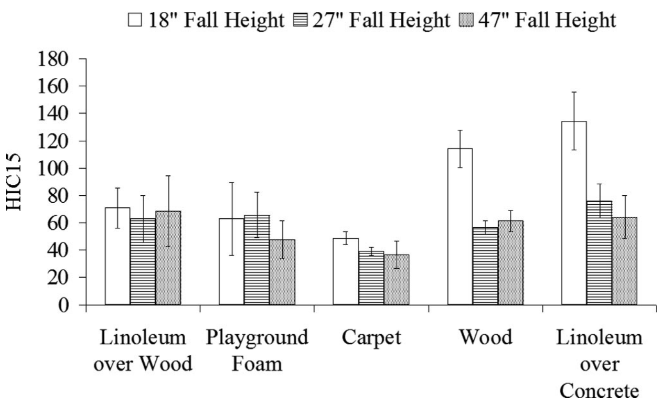

Fig. 4. *Head injury criteria* (HIC_{15}) *for various impact surfaces and fall heights (measured from ground to* ATD *center of mass).* Error bars *represent 95%* CI.

Table 4 Significant Differences in HIC_{15} Values for Various Heights and a Fixed Impact Surface

	18 inches	27 inches	47 inches
Linoleum over wood			
18 inches			
27 inches			
47 inches			
Playground foam			
18 inches			
27 inches			
47 inches			
Carpet			
18 inches			X
27 inches			
47 inches	X		
Wood			
18 inches		X	X
27 inches	X		
47 inches	X		
Linoleum over concrete			
18 inches		X	X
27 inches	X		
47 inches	X		

"X" indicates significant difference ($p < 0.05$).

Table 5 Significant Differences in HIC_{15} Values for Various Surfaces and a Fixed Fall Height

	Linoleum Over Wood	Playground Foam	Carpet	Wood	Linoleum Over Concrete
18 inches					
Linoleum over wood				X	X
Playground foam				X	X
Carpet				X	X
Wood	X	X	X		
Linoleum over concrete	X	X	X		
27 inches					
Linoleum over wood			X		
Playground foam			X		
Carpet	X	X			X
Wood					
Linoleum over concrete			X		
47 inches					
Linoleum over wood			X		
Playground foam					
Carpet	X				
Wood					
Linoleum over concrete					

"X" indicates significant difference ($p < 0.05$).

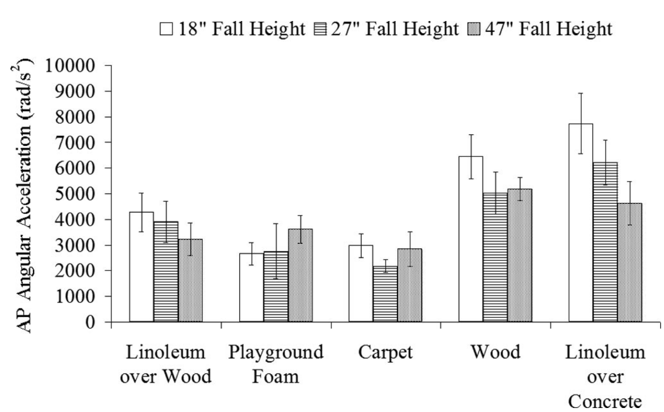

Fig. 5. *Peak anterior-posterior angular head accelerations for various impact surfaces and fall heights (measured from ground to* ATD *center of mass).* Error bars *represent 95%* CI.

18-inch fall height was associated with significantly greater HIC_{15} values than either the 27-inch or 47-inch fall heights (Table 4). For the 18-inch fall height, the wood and linoleum over concrete surfaces were associated with significantly greater HIC_{15} values than other surfaces. For the 27-inch fall height, carpet was associated with significantly lower HIC_{15} values than all other surfaces except wood. For the 47-inch fall experiments, only one significant difference was found across the various surfaces (Table 5).

Angular Head Acceleration

The mean peak AP angular acceleration across all trials was 4246 rad/s^2 (95% CI, 3945–4546) (Fig. 5). The maximum value was 11,804 rad/s^2 (18-inch fall onto linoleum over concrete). Significant differences in angular head acceleration between fall heights occurred only for carpet, wood, and linoleum over concrete surfaces (Table 6). Significant differences between impact surfaces occurred across all fall heights (Table 7).

Impact Duration

The mean head impact duration across all trials was 19.9 milliseconds (95% CI, 19.2–20.6) (Fig. 6). The only significant differences in impact durations due to varying fall height

Table 6 Significant Differences in Anterior-Posterior Angular Head Accelerations for Various Heights and a Fixed Impact Surface

	18 inches	27 inches	47 inches
Linoleum over wood			
18 inches			
27 inches			
47 inches			
Playground foam			
18 inches			
27 inches			
47 inches			
Carpet			
18 inches		X	
27 inches	X		
47 inches			
Wood			
18 inches		X	X
27 inches	X		
47 inches	X		
Linoleum over concrete			
18 inches			X
27 inches			X
47 inches	X	X	

"X" indicates significant difference ($p < 0.05$).

Table 7 Significant Differences in Anterior-Posterior Angular Head Accelerations for Various Surfaces and a Fixed Fall Height

	Linoleum Over Wood	Playground Foam	Carpet	Wood	Linoleum Over Concrete
18 inches					
Linoleum over wood		X		X	X
Playground foam	X			X	X
Carpet				X	X
Wood	X	X	X		
Linoleum over concrete	X	X	X		
27 inches					
Linoleum over wood			X		X
Playground foam				X	X
Carpet	X			X	X
Wood		X	X		
Linoleum over concrete	X	X	X		
47 inches					
Linoleum over wood				X	X
Playground foam				X	
Carpet				X	X
Wood	X	X	X		
Linoleum over concrete	X		X		

"X" indicates significant difference ($p < 0.05$).

occurred on the wood and linoleum over concrete surfaces (Table 8). For these two surfaces, the 18-inch height was associated with shorter durations than the 27-inch or 47-inch heights. Significant differences for varying impact surface occurred across all fall heights (Table 9). Carpet and playground foam were associated with significantly longer durations than all other surfaces.

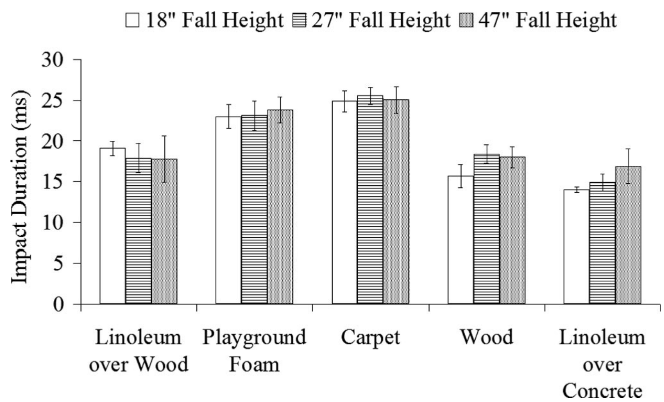

Fig. 6. *Head impact durations for various impact surfaces and fall heights (measured from ground to* ATD *center of mass).* Error bars *represent 95%* CI.

Table 8 Significant Differences in Head Impact Durations for Various Heights and a Fixed Impact Surface

	18 inches	27 inches	47 inches
Linoleum over wood			
18 inches			
27 inches			
47 inches			
Playground foam			
18 inches			
27 inches			
47 inches			
Carpet			
18 inches			
27 inches			
47 inches			
Wood			
18 inches		X	X
27 inches	X		
47 inches	X		
Linoleum over concrete			
18 inches			X
27 inches			
47 inches	X		

"X" indicates significant difference ($p < 0.05$).

DISCUSSION

For the fall scenarios tested, there were few differences in head injury outcome measures associated with varying fall height. Where significant differences did occur, the lowest fall height (18 inches) was typically associated with greater linear and AP angular head accelerations and shorter head impact durations than the two greater fall heights. This sug-

Table 9 Significant Differences in Head Impact Durations for Various Surfaces and a Fixed Fall Height

	Linoleum Over Wood	Playground Foam	Carpet	Wood	Linoleum Over Concrete
18 inches					
Linoleum over wood		X	X	X	X
Playground foam	X			X	X
Carpet	X			X	X
Wood	X	X	X		
Linoleum over concrete	X	X	X		
27 inches					
Linoleum over wood		X	X		X
Playground foam	X			X	X
Carpet	X			X	X
Wood		X	X		X
Linoleum over concrete	X	X	X	X	
47 inches					
Linoleum over wood		X	X		
Playground foam	X			X	X
Carpet	X			X	X
Wood		X	X		
Linoleum over concrete		X	X		

"X" indicates significant difference ($p < 0.05$).

gests an increased injury risk for the 18-inch fall height, a ground-based fall, compared with the two greater fall heights (27 inches and 47 inches). Conversely, similar studies report an increase in head injury risk with increasing fall height.[13,14] The differences can likely be explained by the fall dynamics and, in part, by the initial (prefall) position. The initial position in this study investigated a feet-first fall, where as a previous study by Prange et al. which showed increasing head injury risk with increasing fall height, investigated falls from a supine position with the head in a leading position. In the 18-inch ground-based falls, from a fall dynamics perspective, the ATD rotated rearward about the feet after release and the first major impact with the ground surface occurred at the pelvis followed by the torso and head. The fall dynamics in the 27-inch and 47-inch falls were similar to each other, and the only significant difference between these two was found for angular head acceleration in falls onto linoleum over concrete. In falls from these two elevated heights, when the feet impacted the ground, hip and knee flexion followed and a large portion of the energy from the fall was absorbed in the legs. Additionally, the greater head accelerations associated with the 18-inch fall height occurred to the largest extent in falls onto wood and linoleum over concrete surfaces. This could possibly be explained by differences in surface properties and their effect on fall dynamics. That is, the wood and linoleum over concrete surfaces had the lowest COF. In the 27-inch and 47-inch falls onto these surfaces, the feet tended to slip upon impact causing a second impact at the buttocks

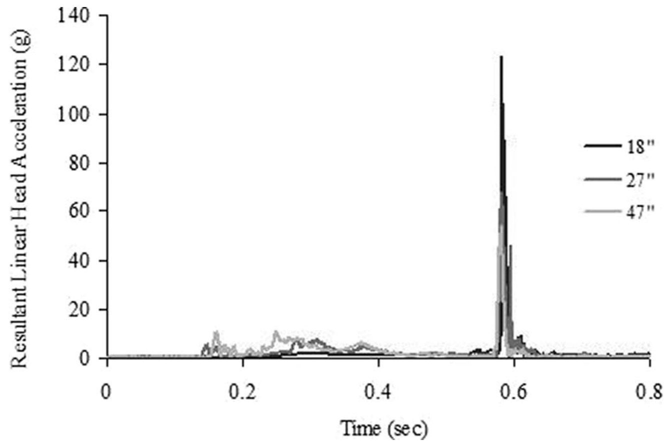

Fig. 7. *Representative linear head accelerations for 18 inches, 27 inches, and 47 inches falls onto linoleum over concrete.*

after the initial foot impact. The second impact at the buttocks absorbed a portion of the energy from the fall, so there was less energy available at head impact leading to lower head accelerations in these falls as compared with the 18 inch falls. For falls in which minimal or no slipping occurred, the ATD rotated rearward about the feet, and although there was generally an impact at the buttocks before head impact, it was not as severe, and the ATD head maintained its momentum. This difference in fall dynamics because of foot slipping is illustrated in Figure 7. It can be seen that from approximately 0.1 seconds to 0.4 seconds, the 27-inch and 47-inch falls generated secondary peaks that did not occur in the 18-inch fall. These secondary peaks represent buttocks impact and an associated reduction in fall energy.

Surfaces with lower COR (less resilient) were associated with shorter impact durations. The magnitudes of the accelerations are dependent on the duration of the impact. For longer durations, the impact is spread over a longer period of time, so lower accelerations result. Likewise, for a given event, shorter durations result in greater accelerations. Because greater head accelerations are associated with a greater injury risk, surfaces with lower COR (linoleum over concrete, linoleum over wood, and wood) can be expected to be associated with increased injury risk, and surfaces with the highest COR (playground foam and carpet) can be expected to be associated with reduced injury risk, which is consistent with the results of this study. These findings are consistent with other studies that examined surface effects on injury risk.[13–15] The surface with the lowest COR in this study was linoleum over wood. Based on the COR measurement alone, one would expect this surface to be associated with the greatest head accelerations. However, the wood and linoleum over concrete surfaces often were associated with significantly greater accelerations than the linoleum over wood surface. This is likely due to both the wood and linoleum over concrete surfaces having lower COF values which affected the fall dynamics. This illustrates that the COF and COR surface properties should be considered together when predicting head injury risk.

HIC values are based on linear head acceleration and time exposure, and can be used to predict the probability of contact-type head injuries. The HIC was developed for use in the automotive industry to assess head injury risk in motor vehicle crash testing, but it has also been used to assess head injury risk in falls, particularly in the playground safety area.[15,16] The proposed HIC_{15} limit for the CRABI 12-month-old ATD is 390. For this limit, there is a 31% probability of skull fracture.[12] All HIC_{15} values measured in these experiments were 173 or less. Therefore, contact-type head injuries would not be expected in these falls.

A large range of linear head acceleration tolerance limits have been proposed. Sturz[17] proposed tolerance limits for 6- to 7-year-old children based on accident reconstruction. An acceleration of 83*g* was proposed as the level above which AIS 2+ injuries could occur. Mohan et al.[18] reported an injury tolerance limit of 200*g* to 250*g* peak acceleration for children (ages 1 to 10) based on a study of head-first free falls. Cory et al.[19] reported several tolerance limits ranging from 50*g* to 200*g* for children (age not specified), where 50*g* is the maximum before-injury threshold and 200*g* is the threshold for fatal injury. Our maximum linear head acceleration across all falls was 130*g* (18-inch fall onto the linoleum-tiled concrete surface). The results of this study were all below the tolerance of 200*g* proposed by Mohan et al. All of the tested scenarios except the 18-inch falls onto linoleum over concrete produced mean peak linear head accelerations below Sturtz's proposed limit of 83*g*. However, the maximum for some fall trials onto linoleum over concrete (all heights), wood (18-inch fall height), and linoleum over wood (27-inch and 47-inch fall heights) exceeded Sturtz's proposed limit of 83*g* indicating that AIS 2+ head injuries could be possible in feet-first free falls depending on the impact surface. However, this tolerance limit was developed for children older than the population addressed in this study.

As with the tolerance limits for linear head acceleration, many different angular head acceleration tolerance limits have been suggested, and are often specific to injury type. Ommaya et al.[4] proposed angular head acceleration thresholds of nearly 30,000 rad/s^2 for an infant and nearly 18,000 rad/s^2 for a young child for mild diffuse axonal injury (DAI). Duhaime et al.[9] compared accelerations from results of surrogate shaking and impact studies to thresholds of approximately 35,000 rad/s^2 and 40,000 rad/s^2 for subdural hematoma (SDH) and DAI, respectively, for an infant. Depreitere et al.[20] proposed a tolerance level of approximately 10,000 rad/s^2 for pulse durations less than 10 milliseconds for SDH based on adult cadaver head impact tests. Reported thresholds for concussion range from 4500 rad/s^2 for an adult to 10,000 rad/s^2 for an infant.[4] All test scenarios except for 47-inch falls onto linoleum over wood, 18-inch and 47-inch falls onto playground foam, and 27-inch and 47-inch falls onto carpet included trials with angular accelerations exceeding 4500 rad/s^2. However, only one fall in this study produced angular acceleration levels exceeding the 10,000 rad/s^2 threshold for concussion in an infant, and that was an 18-inch fall onto linoleum over concrete (11,804 rad/s^2). This fall also exceeded the Depreitere et al. proposed SDH threshold, but it should be noted that this threshold was developed for the adult population. All other falls produced angular acceleration levels below the previously mentioned SDH and DAI injury threshold levels. Although the results of this study should not be directly compared with thresholds developed for adults, there is some disagreement in the literature as to how the thresholds would be affected if scaled to the pediatric population. Although some studies have suggested that the thresholds should increase with decreasing brain size,[21] the pediatric skull structure and material properties are very different from that of an adult leading others to suggest that the pediatric brain is much more susceptible to injury.[22,23] Because of this controversy, we opted not to attempt scaling from experimentally derived thresholds.

The low risk of severe head injury seen in this study is consistent with the findings of other studies. In studies evaluating children falling three stories or less, no fatalities were reported (when false histories were ruled out as a possibility).[24–26] Furthermore, Snyder[24] reported no injuries for feet-first free falls from less than 25 ft, even onto a concrete surface. Chadwick et al.[27] reported seven fatalities from falls less than four ft, but concluded that these were likely false histories. Severe head injuries were present in each of the seven cases. Two studies of bed falls found no serious head injuries in a combined 512 cases.[28,29] There were four skull fractures reported in these studies but all were of a nonserious nature.

A study by Prange et al.[13] performed short-distance fall experiments with an anthropomorphic surrogate of a 1.5-month-old infant. The surrogate was initially positioned horizontally with the head slightly below the body such that the head would impact the ground first. For similar heights and surfaces, the angular head accelerations measured by Prange et al. were more than 10 times those measured in this study. The primary explanation for this difference is likely the different initial positions of the surrogates across the two studies. Larger head accelerations would be expected in a head-first impact than in a feet-first impact. Additional differences would also be expected because of the different ages, mass, and skull and neck properties of the surrogates.

A previous study by Bertocci et al.[14] used a 3-year-old ATD in simulations of feet-first falls. The authors reported linear head accelerations up to six times greater than those measured in this study for all similar heights and surfaces. Angular accelerations were not reported. Because the fall kinematics were similar to those observed in this study, the differences in the results are likely because of the differences in age representation and associated characteristics of the ATDs.

Several studies have used animals and human cadavers to determine loads necessary to produce head injuries. Ommaya et al.[4] describes a study by Weber in which 15 infant cadavers were dropped from a height of 32 inches (82 cm) onto stone, tile, carpet, and linoleum covered surfaces. All of the drops were from a horizontal initial position, which allowed the head to impact first, and each drop produced skull fractures. Nahum and Smith[30] simulated impacts to the frontal skull bone of 10 adult cadavers. Accelerations greater than 195*g* (1,910 m/s^2) were associated with head injuries ranging from minor contusions to more severe injuries including subdural hematomas. Ommaya and Hirsch[21] and Gennarelli et al.[31] studied head injuries in primates. The former study applied rotational loads to three primate species and found that smaller loads produced concussion in animals with a larger brain mass. For chimpanzees (which have the largest brain mass of the three primates tested) the onset level of concussion in whiplash occurred at an angular velocity of 70 rad/s^2. For impact duration of 20 milliseconds, this corresponds to an angular acceleration level of 3,500 rad/s^2. In the study by Gennarelli et al., pure rotational loads, ranging from 100,000 to 200,000 rad/s^2, were applied to the heads of rhesus monkeys and baboons. The resulting head injuries ranged from mild concussion to severe DAI. The results of the present study were below the 195*g* level found by Nahum and Smith to produce injury and the loads applied to the primates in the Gennarelli et al. study. Several fall trials exceeded the 3,500 rad/s^2 level found to produce concussion in chimpanzees. However, it could be misleading to compare the results of the cadaver and primate studies with those of the present study. The Ommaya and Hirsch study showed that different load levels are necessary to cause head injury in subjects of different sizes. Therefore the outcomes found in the adult cadaver and primate studies cannot be compared directly with those sustained by a 12-month-old child. Although the chimpanzee has the closest brain mass to that of a 12-month-old, the chimpanzee skull and brain properties, and structure are very different from that of a child. Comparing the Weber study using infant cadavers can also be misleading because Weber used an initial position of the cadaver that is different from that used in this study. Therefore, different head injury outcome measures and a different head injury risk would be expected.

There are several limitations of this study in both the experiments and the thresholds used to determine injury risk. One limitation is the biofidelity of the ATD. The CRABI ATD was designed for use in automotive crash testing and not necessarily for low-energy events such as falls. In particular, the neck appears stiffer than would be expected for a 12-month-old child. Also, the ATD was designed for frontal impact testing so there is little neck rotation out of the sagittal plane. A more flexible neck would likely be associated with increased head accelerations. Therefore, the head accelerations resulting from experiments with the CRABI 12-month-old ATD may be underestimated compared with those that would be experienced by a 12-month-old child. A surrogate with a more biofidelic neck is needed to more accurately assess head injury risk in fall scenarios. In addition to the neck, joints (shoulders, elbows, hips, and knees) of the ATD are limited to motion primarily in the sagittal plane. Although the evaluated falls occurred primarily in the sagittal plane, there was out of plane motion in several falls, which may lead to inaccuracies in kinematics and head acceleration measures. There is also no active muscle response in the ATD, but this likely has little effect on the accuracy of the results because active motion during a fall in a 12-month-old child would be minimal. Despite these limitations, the CRABI ATD is a reasonable anthropomorphic surrogate of a 12-month-old child and was designed with particular attention to head and neck biofidelty in high-speed impact. Although use of this ATD to predict injury risk in low energy events was not the original intent of the ATD design, several studies have used ATDs to gain an understanding of injury risk in falls.[14,32,33] A few studies have also developed customized pediatric surrogates to investigate injury risk in low-energy events.[9,13,34] Additionally, a study by Prange et al.[35] compared the head impact response of the CRABI 6-month-old ATD with that of pediatric cadaveric head specimens in drop tests and found the results to be comparable for vertex, occiput, and forehead impacts. The comparable head impact response found across the CRABI ATD and pediatric cadaver head specimens supports the utility of the CRABI ATD in fall-related experiments investigating head injury. The authors think that use of the CRABI 12-month-old ATD provides an important first step in understanding complex fall dynamics and their interaction with the fall environment.

In addition to limitations in the ATD biofidelity, the ATD was not equipped with footwear in these fall experiments. It is likely that COF values would be different if the ATD was equipped with shoes or socks, which in turn could affect the fall dynamics and resulting outcome measures.

Another limitation of this study concerns the injury criteria used to predict injury risk. Little information exists on injury tolerances in children. Accordingly, proposed criteria (including those presented in this paper) have typically been scaled from either adult data or from primate studies. Scaling generally accounts for mass differences, but may also attempt to account for differences in geometry and material properties. Pediatric HIC thresholds are based on scaling from adult data using mass, geometry, and material properties.[12] However, because there is limited information on pediatric brain tissue properties, the material properties of brain tissue were assumed to have the same age variation as calcaneal tendon.[12] In addition to this assumption, the thresholds may be inadequate because they do not account for the different structure of a child's skull compared with an adult skull. A 12-month-old child's skull has several flexible fontanelles and sutures that allow for greater skull deformation than that of an adult under the same load. This also allows for more brain deformation. In a study of infant skull and suture properties investigating loading at rates similar to those that would occur in short falls, it was found that pediatric suture deforms 30 times more than pediatric cranial bone and 243 times more than adult cranial bone.[22] Also, brain tissue properties have been found to be age dependent.[23,36] Thibault and Margulies[23] applied scaling based on brain tissue properties to angular acceleration thresholds for concussion, subdural hematoma, and DAI originally derived from brain mass scaling alone, and found that the injury thresholds were reduced. These differences between the adult and infant skull and brain properties highlight the need for more accurate pediatric head injury criteria.

CONCLUSIONS

This study examined the risk of severe head injuries in short-distance feet-first free falls using an ATD representing a 12-month-old child. Differences in head injury risk because of changing fall environment characteristics (fall height, impact surface) were also assessed. Ground-based falls were associated with an increased head injury risk as compared with the higher fall heights tested. Head injury risk also increased for surfaces with lower COR. However, the risk of severe head injury for a 12-month-old child in feet-first free falls across all tested surfaces and heights was low. The results of this study can provide a first step in aiding clinicians in determining whether a child's injuries are consistent with the stated cause, and highlight the importance of obtaining a detailed history that includes initial position, fall height, impact surface properties, and fall dynamics when evaluating head injury risk for a short-distance fall.

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