



7-2010

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Suggested Citation:

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<http://thejns.org/doi/pdf/10.3171/2010.3.PEDS09357>

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Abstract

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Methods. To identify potential age-dependent mechanical load factors that may be responsible for these clinical findings, the authors created an instrumented dummy representing an 18-month-old child using published toddler anthropometry and mechanical properties of the skull and neck, and they measured peak angular acceleration during low-height falls (1, 2, and 3 ft) onto carpet pad and concrete. They compared these results from occiput-first impacts to previously obtained values measured in a 6-week-old infant dummy.

Results. Peak angular acceleration of the toddler dummy head was largest in the sagittal and horizontal directions and increased significantly (around 2-fold) with fall height between 1 and 2 ft. Impacts onto concrete produced larger peak angular accelerations and smaller impact durations than those onto carpet pad. When compared with previously measured infant drops, toddler head accelerations were more than double those of the infant from the same height onto the same surface, likely contributing to the higher incidence of loss of consciousness reported in toddlers. Furthermore, the toddler impact forces were larger than those in the infant, but because of the thicker toddler skull, the risk of skull fracture from low-height falls is likely lower in toddlers compared with infants.

Conclusions. If similar fracture limits and brain tissue injury thresholds between infants and toddlers are assumed, it is expected that for impact events, the toddler is likely less vulnerable to skull fracture but more vulnerable to neurological impairment compared with the infant.

Disciplines

Biomedical Engineering and Bioengineering | Engineering

Comments

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Biomechanics of the toddler head during low-height falls: an anthropomorphic dummy analysis

Laboratory investigation

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KEY WORDS • anthropomorphic surrogate • low-height fall • toddler • head impact • pediatric head injury

FALLS are the most common environmental setting for closed head injuries treated in pediatric observational units²⁶ and are responsible for 135 in every 100,000 deaths in children 15–17 months of age.² Despite the large body of epidemiological work published on falls and injury outcomes,^{9–11,18,23–25,30,32,36–40,43,45,49,56,58,61,66,67,71,74–76} these studies are limited by the incidence of reported accidents, patient data availability, details of the events, and age distributions, and cannot provide information about the biomechanics of a given injury event. Clinical studies of children injured from falls also present a large heterogeneity of resulting injuries and suggest that

fall height and impact surface contribute to injury severity. However, the lack of agreement on critical fall height and impact conditions across these studies highlights the need for a more controlled environment in which to assess fall conditions contributing to head injury severity from childhood falls.

Anthropomorphic test dummies can be used to measure response corridors in a controlled setting with the goal of understanding the kinematics of an event for diagnosis and/or prevention of injuries. Existing pediatric surrogates such as the anthropomorphic Hybrid III (First Technology Safety Systems) child dummies and CRABI (First Technology Safety Systems) dummies have provided a wealth of information about body forces and injury severity during motor vehicle accidents^{31,65,68,70} and fall simulations.^{7,8,16} The Hybrid III and CRABI dummies have been used to establish HIC tolerance levels for

Abbreviations used in this paper: CRABI = child restraint air bag interaction; HIC = head injury criterion; HIC₁₅ = HIC with a maximum time interval of 15 msec; HIC₃₆ = HIC with a maximum time interval of 36 msec.

high head acceleration scenarios in children.^{21,33} However, these existing anthropomorphic test dummies assume the dummy head to be a rigid body in which injury acceleration thresholds are scaled from adult values based on the assumption that mass and skull material differences vary by relatively simple mathematical relationships.^{20,29} Additionally, surrogates such as the CRABI are designed for use in child restraint systems in which the child does not experience a hard surface head impact and therefore may not give a “childlike response” for head impact scenarios.²⁹ In sum, commercial child anthropomorphic test dummies lack the appropriate head properties of infants and toddlers which, because they are different from adults, affects the biofidelity of the impact response.^{13,27,35,41,42}

Our laboratory recently developed a biofidelic anthropomorphic surrogate for the 6-week-old infant (with no Hybrid equivalent) to investigate pediatric head injury and abusive head trauma. The current study builds on this infant work with the development of a surrogate for the 18-month-old toddler to investigate head accelerations in an age group that commonly experiences head injuries from low-height falls. In the toddler surrogate, body weight, body length, and neck stiffness were increased relative to the infant surrogate, as documented with development,^{5,6,46,60} skull thickness was nearly doubled,¹ and a fused skull was used to simulate closed sutures and fontanelles.^{3,13,78}

In the current study we will characterize the kinematic head response of the toddler during drop tests from 1, 2, and 3 ft onto 2 surfaces (carpet pad and concrete). We expect that stiffer surfaces and higher heights will produce larger head acceleration. Finally, we will compare the toddler head response to the previously measured infant head response for the same height-surface combinations.

Methods

We developed a novel biofidelic 18-month-old toddler surrogate using published anthropometry^{29,69,72} and mechanical properties of the neck (Fig. 1).^{46,47,60} The head of a toy doll was adjusted using lead shot to model the appropriate head weight. The neck and head were altered as described in the next sections. The total head mass, head circumference, head height, and neck length were 2.32 kg, 45.1 cm, 17.1 cm, and 3.9 cm, respectively, and compared well with average values of 50th percentile 18-month-old toddlers.^{29,69,72}

Neck

Currently, there are no published data on the flexion and extension bending stiffness of the human toddler cervical spine. We calculated a bending stiffness corridor for the human 18-month-old toddler cervical spine based on previously published data in adults and pediatric caprine and primate models.^{28,44,46,55,60} Taken together, the human, caprine and primate data yielded estimates for human toddler total cervical spine stiffness corridors of 0.036–0.102 Nm/° in flexion and 0.039–0.07 Nm/° in extension. To simplify the surrogate neck, we chose to model the toddler neck stiffness the same in flexion and extension. The neck, constructed from dryer tubes and nylon, had a

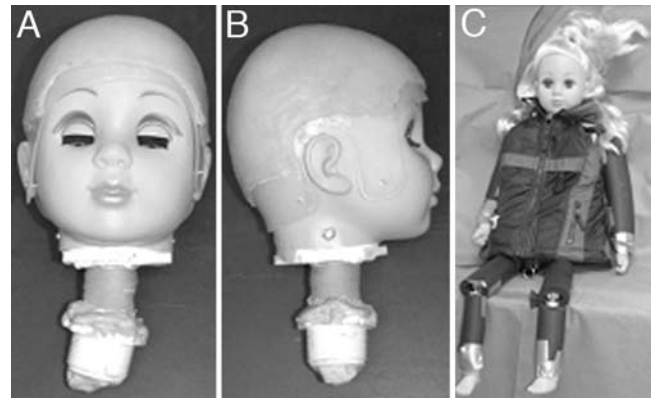


FIG. 1. Photographs. Skull casing (A) and scalp (B) of the 18-month-old surrogate. The total body is shown (C).

total flexion/extension stiffness of 0.0637 Nm/°, which is in within our target toddler neck stiffness corridors and also the 95% CI for human pediatric cervical spine.⁵⁵

Skull

The skull was constructed of a biofidelic copolymer material. Although there are no published data for the elastic modulus of the toddler skull, we measured the elastic modulus of a single sample obtained from the occipital bone of a 36-month-old human toddler (protocol approved by the international review boards of the University of Pennsylvania and Children’s Hospital of Philadelphia). We found that the elastic modulus (321 MPa) did not differ (within the 95% CI) from the infant (0–1 year, 329 ± 55.3 MPa).¹³ We therefore modeled the skull using the same copolymer (elastic modulus 535 ± 138.8 MPa) used previously in an infant anthropomorphic surrogate¹⁴ but increased the skull thickness from the infant value of 1.5 mm to 2.5 mm.³⁴ The geometry of the skull (head circumference and head height) matched those of a human toddler (Table 1). Similarly, a rubber material (elastic modulus 1.20 ± 0.05 MPa)^{62,63} previously shown to have scalplike material properties (elastic modulus 1.54 MPa)²² was overlaid on the cranium to mimic the scalp layer.

Torso and Appendages

The torso, arms, and legs were constructed from aluminum and enclosed in foam padding to simulate the compliance of soft tissue, and they were also weight-adjusted to appropriate toddler values reported in the literature (Table 1). The total weight of the doll was 11 kg to match the body weight of a 50th percentile 18-month-old toddler.

Testing Protocol

A custom-built 9-accelerometer (7264B-2000, Endevco, Inc.) array was placed inside the head at the center of gravity via a lightweight mounting plate to measure angular accelerations of the head in the sagittal, coronal, and horizontal directions. An angular velocity transducer (ARS-06, Applied Technology Associates) was fixed to the same mounting plate to measure rotational velocity in the sagittal direction. A separate 3-accelerometer array

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TABLE 1: Comparison between body measurements (dimensions and mass) in an 18-month-old toddler and the surrogate*

Body Measurement	Value	
	18-Mo-Old Toddler	Surrogate
head		
circumference (cm)	47.8†	45.1
height (cm)	16.95†	17.1
mass (kg)	2.67‡	2.32
neck		
length (cm)	3.8†	3.9
ext stiffness (Nm/°)	0.039–0.062§	0.064
flex stiffness (Nm/°)	0.036–0.102§	0.064
torso		
length (cm)	29.2¶	27.0
breadth (cm)	16.2¶	18.5
depth (cm)	7.2¶	7.4
mass (kg)	5.13‡	5.0
arms		
length (cm)	37.7**	36.05
mass (kg)	1.29‡	1.35
legs		
length (cm)	39.3**	37.2
mass (kg)	1.57‡	1.56
total		
height (cm)	80.9**	82.5
mass (kg)	11**	10.55

* ext = extension; flex = flexion.

† As reported in the study by Schneider et al.

‡ As reported in the study by Irwin and Mertz.

§ As reported in the studies by Nightingale et al., Nuckley et al., Ouyang et al., and Pintar et al.

¶ As reported in the study by Snyder et al.

** As reported in the study by Adeloje et al.

was placed in the torso to measure linear acceleration of the body. In this study, emphasis was placed on measuring the angular rather than linear translational motion of the head, as numerous investigators have reported that traumatic head injury is more closely associated with angular rather than translational head accelerations.^{50,52,53,57} Furthermore, angular acceleration of the head has been shown to cause stretching and shearing of the underlying vascular and white matter tissue, which are responsible for common clinical manifestations of pediatric traumatic brain injury such as subarachnoid hemorrhage and diffuse axonal injury.^{50,52,53}

The toddler surrogate was subjected to a series of free-fall drop tests onto a 0.25-in household carpet pad and concrete. The limbs were restrained over the surrogate's chest to prevent as much interference as possible during a drop. For each drop test the doll was oriented in the supine position with the head about 15–20° lower than the feet, to ensure that the occiput made contact with the surface before the torso. This orientation of the body and head simulated a near-worst case scenario, with maxi-

imum head impact force and rotations after contact. Had the body been oriented in the horizontal position, the initial thoracic contact with the surface would decrease head contact and subsequent rotation. Vertex impact would maximize head impact force and likely cause neck compression, but it would eliminate significant head rotational accelerations. As a result, the measured angular accelerations likely represent maximal head responses. For our occiput-first conditions, the surrogate was dropped from 1, 2, and 3 ft. The drops onto 2 impact surfaces from 3 heights were performed 10 times, for a total of 60 drops. Acceleration and velocity time histories were recorded.

The 9 head acceleration traces from the infant and toddler, 3 body angular velocity traces, and single head angular velocity trace were imported into Matlab (MathWorks, Inc.) and filtered using a fourth order Butterworth low-pass filter with a cutoff frequency of 1650 Hz. For the 9 time histories recorded in the head, average acceleration curves were calculated for each direction using a previously validated optimization program created in Matlab.¹² Briefly, the data were optimized by taking the 3 acceleration traces oriented in the sagittal direction and resolving them into an average acceleration time history for that direction. The same was done for acceleration in the horizontal and axial directions to obtain an average acceleration time history along each of the 3 coordinate axes (x, y, and z). The angular velocity trace recorded in the sagittal direction was differentiated and used to confirm the average angular acceleration in that direction. Peak acceleration in each direction was extracted for statistical analysis. Resultant angular acceleration was calculated by taking the square root of the sum of squares of accelerations in the sagittal, horizontal, and coronal directions.

Impact force was calculated for each drop using the largest peak angular acceleration over all 3 directions and the midlocation of the cervical spine as the center of rotation. Although the center of rotation in a human child may not necessarily be in the center of the cervical spine due to differences in the stiffness of the vertebral motion segments, the surrogate neck is uniform in composition, and thus it is uniform in stiffness along the long axis. Therefore, we chose the center of rotation as the middle of the neck (at approximately C-4). Force was calculated as $F = mr\theta_p$, where m is the mass of the head ($m = 2.32$ kg) and r is the distance from the center of rotation (C-4) to the center of gravity of the head ($r = 0.0755$ m).

The influence of contact surface material and drop height was evaluated using an ANOVA. Directional differences in head acceleration response for each drop condition were also evaluated using an ANOVA. The results from the toddler surrogate drops were compared with previously published infant impact acceleration and contact force from infant surrogate drops from the same heights onto the same surfaces.¹⁴

Results

A total of 60 drops were performed, yielding 53 successful drops from the 3 heights onto 2 surfaces (1 ft onto concrete [9 drops], 1 ft onto carpet pad [9 drops], 2 ft onto concrete [8 drops], 2 ft onto carpet pad [10 drops],

3 ft onto concrete [7 drops], and 3 ft onto carpet pad [10 drops]). We excluded 7 drops because the recorded angular velocity in the sagittal direction did not match well with the velocity calculated from the angular acceleration trace in that direction. For all falls, we confirmed from video recordings that the occipital region of the head made contact with the impact surface first followed by the torso. Accelerometer data from the torso also verified that the torso impact followed head impact in all cases. A typical drop consisted of the initial head impact followed by a rapid deceleration. Surprisingly, little to no rebound was observed in any scenario, indicating that there was no reversal of direction of the head after the rapid deceleration. The maximum peak angular acceleration was defined by evaluating each of the 3 directions separately and by considering both acceleration and deceleration immediately after the initial contact event (Fig. 2). Across all drops, 55% of the maximum peak accelerations occurred during the angular acceleration phase and 45% during the deceleration phase.

Peak Angular Acceleration

For all 6 drop scenarios, peak angular accelerations were significantly lower in the coronal direction compared with the sagittal and horizontal directions ($p < 0.05$, Fig. 3 upper). Because the majority of angular motion was in the sagittal and horizontal directions, we chose to focus on these 2 directions for the remaining statistical analyses. Peak angular acceleration in the sagittal and horizontal directions occurred nearly simultaneously, separated by no more than approximately 0.05 msec. A 2-factorial ANOVA (height and surface) found an increase in height and surface stiffness to significantly increase peak angular acceleration in the sagittal ($p < 0.003$) and horizontal ($p < 0.001$) directions. The interaction between height and surface stiffness had a significant effect on angular acceleration in the horizontal direction ($p < 0.02$). Post hoc analysis showed that in the sagittal direction, peak angular accelerations were

consistently higher for drops onto concrete than for drops onto carpet pad at each height ($p < 0.05$, Fig. 3). Although peak angular acceleration in this direction did not vary with drop height in concrete impacts, carpet pad impacts resulted in significantly lower peak angular accelerations from 1 ft compared with 2 and 3 ft.

Peak-to-Peak Change in Angular Velocity

A similar trend in directionality was observed with angular velocity as with angular acceleration: velocities in coronal plane were smaller than those in the sagittal and horizontal velocities (Fig. 4 upper). A 2-factorial ANOVA (height and surface) found an increase in height to significantly increase peak change in angular velocity in the sagittal ($p < 0.0001$) and horizontal ($p < 0.001$) directions. An increase in surface stiffness significantly increased peak change in angular velocity in the sagittal direction ($p < 0.002$, Fig. 4 lower). The interaction between height and surface had a significant effect on angular velocity in the horizontal direction ($p < 0.03$). In the post hoc analysis, the peak-to-peak change in angular velocity in the sagittal direction was height dependent in both concrete and carpet pad drops, with velocities from 3 ft being larger than 1 and 2 ft ($p < 0.05$, Fig. 4 lower). A surface dependence was noted in drops from 2 ft, but the same trend was not observed at 1 or 3 ft.

Calculated Impact Force

The estimated impact force for each drop scenario, calculated using the maximum peak angular acceleration in any direction, is presented in Fig. 5. A 2-factorial ANOVA (height and surface) found an increase in height and surface stiffness to significantly increase the impact force ($p < 0.0001$). Post hoc analysis showed that concrete drops produced significantly higher impact forces than carpet pad from 1 and 3 ft. On both surfaces, 3-ft drops resulted in higher impact forces than those at 1 ft ($p < 0.05$, Fig. 5).

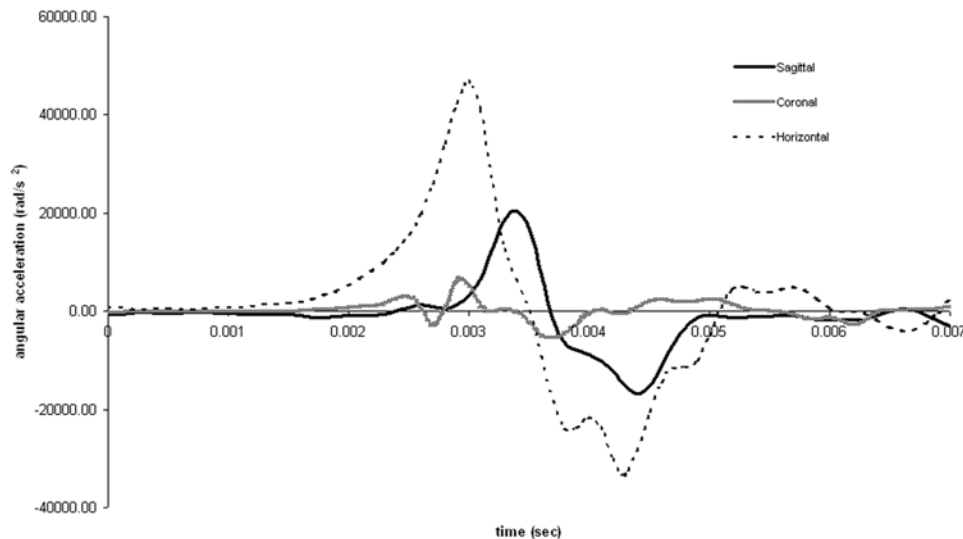


Fig. 2. Graph showing the representative head acceleration trace. Example of angular head acceleration in the sagittal, coronal, and horizontal directions in a drop onto concrete from 1 ft.

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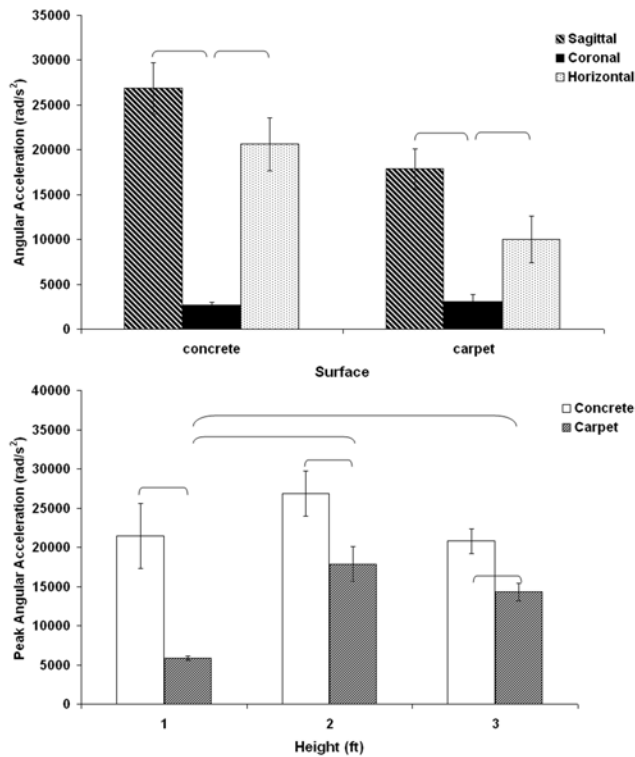


FIG. 3. Peak angular acceleration of the toddler surrogate head. Representative directional differences for concrete drops from 2 ft (**upper**) and height/surface differences in the sagittal direction (**lower**). Brackets indicate groups that are significantly different ($p < 0.05$).

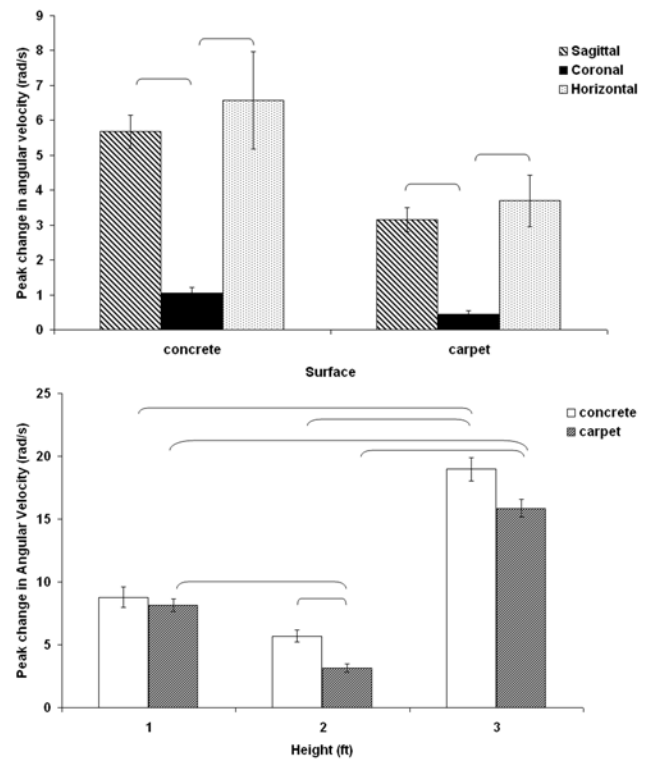


FIG. 4. Peak-to-peak change in angular velocity of the toddler surrogate head. Representative directional differences for concrete drops from 2 ft (**upper**) and height/surface differences in the sagittal direction (**lower**). Brackets indicate groups that are significantly different ($p < 0.05$).

Comparison With Infant

Drop tests in the toddler surrogate were compared with previously reported values for a 6-week-old infant anthropomorphic surrogate¹⁴ (Figs. 6 and 7). In infant and toddler surrogates, the primary head motion was in the sagittal and horizontal directions, with very little motion in the coronal direction. An ANOVA for acceleration showed that peak angular acceleration significantly increased with age in all 3 directions ($p < 0.02$). On average, peak angular accelerations of the head in all 3 directions were 80% larger in the toddler than in the infant (Fig. 6). In the sagittal direction, drops from 2 ft onto either surface and drops from 1 ft onto concrete resulted in significantly higher peak angular acceleration in the toddler dummy compared with the infant ($p < 0.05$). Unlike the infant surrogate, peak accelerations in the toddler were significantly larger for impacts onto concrete than carpet pad in the sagittal direction. Also, toddler drops onto concrete were not significantly affected by height in the sagittal direction while in the infant both concrete and carpet pad exhibited height dependence.

A separate ANOVA for velocity showed that peak-to-peak change in angular velocity was significantly affected by age in all 3 directions ($p < 0.0001$). Over all 3 directions, peak-to-peak change in angular velocity was approximately 85% smaller in the toddler than in the infant. Peak-to-peak change in angular velocity increased with increasing heights in both the infant and the toddler onto both concrete and carpet pad. In the sagittal direc-

tion, angular velocity was significantly lower in the toddler compared with the infant ($p < 0.05$) from all heights onto all surfaces.

Finally, average impact force was significantly larger in the toddler than in the infant ($p < 0.0001$) at all heights onto all surfaces (Fig. 7). Although the impact force in the toddler did not appear to change with height, the impact force in the infant increased with increasing height. A similar dependence on surface was observed in both ages at 2 and 3 ft with concrete producing higher impact

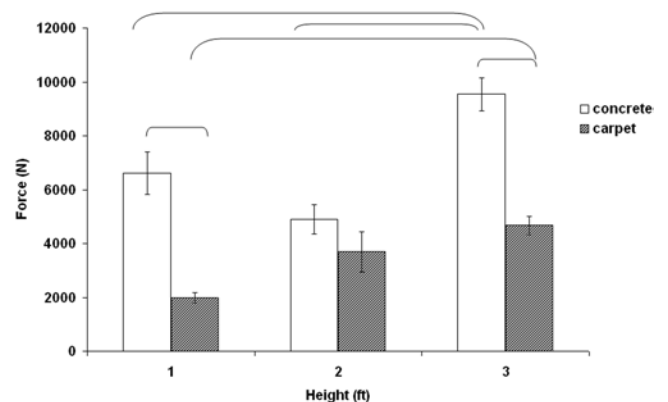


FIG. 5. Estimated impact force of the toddler surrogate head for 3 heights and 2 surfaces using a center of rotation in the middle of the neck. Brackets indicate groups that are significantly different ($p < 0.05$).

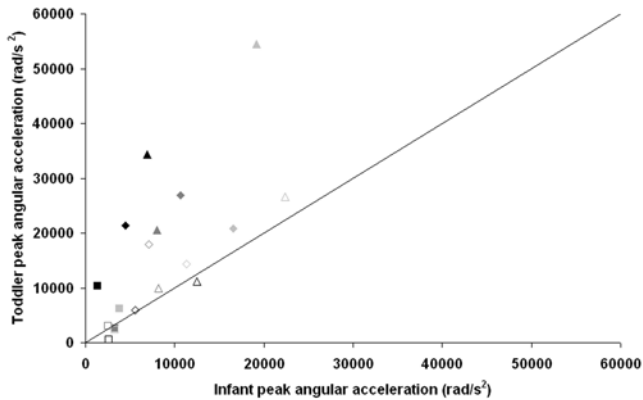


Fig. 6. Toddler versus infant peak angular acceleration. Peak angular acceleration was significantly higher in the toddler compared with the infant ($p < 0.05$). This was more pronounced in drops onto concrete drops (closed symbols) than in drops onto carpet (open symbols). Black, dark gray, and light gray represent drops from 1, 2, and 3 ft, respectively. Diamonds, rectangles, and triangles represent the mean peak sagittal, coronal, and horizontal accelerations, respectively.

forces. This surface modulation was also observed in the toddler at 1 ft but not in the infant.

Discussion

Falls are the most common cause of head injury in children 1–4 years of age,⁴ but currently there are limited data for toddler head kinematics during an impact event. Previous data have been published for feet-first free-fall events and simulated falls from a bed or couch using the Hybrid II 3-year-old dummy.^{7,8,16} However, because there are little to no data defining the material properties of the toddler skull, the Hybrid II toddler dummy was developed using scaled adult values, and it may overestimate material properties of the skull and scalp, affecting the biofidelity of the dummy response. The current study adds novel data for head response in the toddler using an anthropomorphic dummy with a biofidelic head.

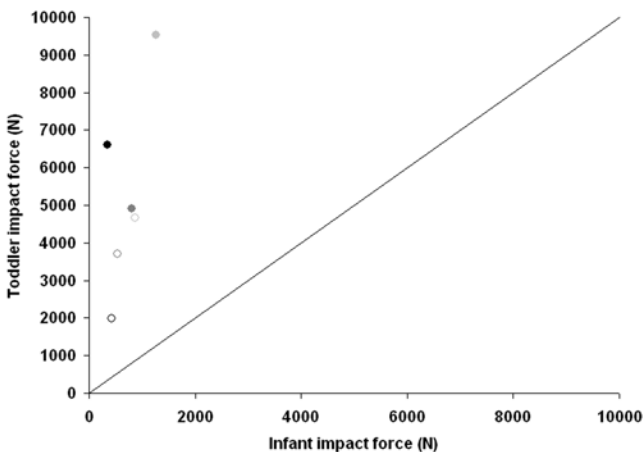


Fig. 7. Toddler versus infant impact force. Impact force was significantly higher in the toddler than in the infant at all heights onto all surfaces except 1 ft onto carpet pad ($p < 0.05$). Black, dark gray, and light gray represent drops from 1, 2, and 3 ft, respectively. Closed and open shapes represent concrete and carpet pad drops, respectively.

In drops from 1, 2, and 3 ft onto carpet pad and concrete, the majority of head motion occurred in the sagittal and horizontal directions. These predominant planes of motion suggest that on contact with the surface the head mostly rebounds up, perpendicular to the impact surface and rotates horizontally, creating horizontal acceleration. Horizontal rotations of the head were observed in video recordings of the drops and are most likely attributed to the rounded geometry of the head, creating an instability at the impact site. Once it began turning, the head would tend to continue rotating about the z axis until the ear or cheek made contact with the surface, resulting in a more stable head position. Very little rotation was observed in the coronal direction, indicating that lateral (shoulder-to-shoulder) motion was minimal.

Interestingly, we found that height only influenced head acceleration in drops onto carpet pad from ≤ 2 ft. No differences in sagittal head acceleration were observed between 2- and 3-ft drops on carpet pad. The 0.25-in-thick carpet pad used in this study represents the typical carpet pad in household settings. Our data suggest that for drops from 1 ft, the carpet pad absorbs enough energy during impact to reduce head acceleration in the sagittal direction. Likely at higher heights, the carpet pad compresses and provides some energy absorption compared with concrete alone (because drops onto concrete have still higher accelerations), but not enough for the height of the drop to significantly influence head rebound in the sagittal direction. No differences in head acceleration across height were observed in drops onto concrete, but concrete drops resulted in significantly higher head accelerations compared with carpet pad at all heights. We also observed longer time durations in drops from 3 ft compared with 2 ft onto carpet and compared with 1 and 2 ft onto concrete. We suspect that the longer time duration in drops from 3 ft may be attributed to deformation of the copolymer skull on impact. Assuming the copolymer at the occiput can be approximated as a simply supported plate under a concentrated load at the center,⁷³ we estimate, based on the impact force (2–9.5 kN) and the material properties (elastic modulus 535 MPa) of the copolymer, that the maximum deflection of the copolymer in drops from 3 ft onto concrete (approximately 2 mm) is 4.75 times higher than the deflection in drops from 1 ft onto carpet pad, which likely results in longer impact duration times.

Our results agree with previous studies that showed that head kinematics may be influenced by impact surfaces.^{7,14–16,63} Cory and Jones¹⁵ developed a simulation system for testing potential severity of head impacts onto playground and household surfaces. A surrogate head-form, representing the pediatric head, was attached to a drop tower and released onto various surfaces including concrete, carpet, wood chips, and linoleum. Concrete was found to produce the largest linear acceleration compared with all other surfaces tested. Bertocci et al.⁷ also noted a dependence on impact surface when feet-first free falls were simulated using the Hybrid II 3-year-old dummy. Although concrete was not tested in the Bertocci study, playground foam was found to produce significantly lower peak linear head acceleration than carpet, linoleum,

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and wood.⁷ In a subsequent study from the same group using the same dummy, dry surfaces were associated with higher linear head acceleration and HIC values than wet surfaces.¹⁶ While these previous studies showed a relationship with impact surface and linear head acceleration, they did not report values for angular head acceleration, which is shown to be associated with traumatic brain injury.^{50–52} Others have reported angular head acceleration for 6-week-old infant anthropomorphic surrogates, and they have also shown increasing acceleration with increasing surface stiffness.^{14,63} Our results improve on these previous studies by measuring angular acceleration in 3 directions in a toddler anthropomorphic dummy and by showing that peak angular acceleration also changes with impact surface characteristics.

We noted significant increases in the overall peak angular head acceleration and impact force and decreases in peak-to-peak change in angular velocity and time duration in the 18-month-old toddler surrogate compared with our previously published data from a 6-week-old infant surrogate subjected to the same drop conditions. Several age-appropriate biomechanical differences between the 2 surrogates may account for variations in the kinematic head response. First, the head mass of the toddler is more than twice that of the infant (2.32 vs 1 kg), which contributes to the larger head accelerations and larger impact forces. Second, the total overall body mass of the toddler is also more than twice that of the infant. Because the torso accounts for a large portion of the overall body mass, the mass of the torso significantly influences the motion of the body after impact. From digital video, it was noted in the infant drops that although the legs of the surrogate weighed down the distal torso and prevented it from rotating toward the head of the surrogate, the body of the infant surrogate did move upward in a translational manner following torso impact. In contrast, no such torso translational rebound was observed in the toddler surrogate because the torso weighed itself down. The collision between the toddler torso and the impact surface was likely an inelastic collision and therefore the kinetic energy of the torso just before impact was likely transformed into heat or sound energy on impact.

Third, there is a moderate amount of cyclic head rebound that occurs in the infant drops that is not observed in the toddler, which we attribute to differences in neck stiffness between the infant and toddler. Neck stiffness differed significantly between the 2 surrogates and contributed to the overall unique head kinematic response. The toddler neck is an order of magnitude stiffer in flexion (0.064 vs 0.0059 Nm/°) and 4 times stiffer in extension (0.064 vs 0.0159 Nm/°) than in the infant.¹⁴ The overall stiffer toddler neck may have restrained rebound of the toddler head, whereas the flexible infant neck may not have resisted rebound. Also, the more than 2-fold larger combined mass of the head, neck, and torso (10.55 kg) may dominate the impact event in the toddler, causing the entire dummy to come to rest quickly, whereas the smaller mass of the infant (4.4 kg) allows for more rebound up off the surface following impact. Nevertheless, we hypothesize that neck properties play a role in motion following impact. Anthropomorphic dummy studies that

compare head impact kinematics over a range of neck flexion/extension stiffness are warranted to investigate the specific role that the neck plays in head motion following impact.

Another, albeit more minor, factor that may contribute to the kinematic differences between the infant and toddler is that the toddler surrogate has a fused copolymer skull with 2.5-mm thickness while the infant skull is composed of 5 copolymer plates of 1.5-mm thickness connected by silicone membrane “sutures.” Although both skulls are made from the same copolymer, the thickness and connectivity of the toddler skull prevent large deformations of the skull as a whole. Distortion of the infant skull occurs more readily on impact with carpet or concrete. We attribute the rebound to the more elastic collision of the infant head with the impact surface that likely occurs because of the silicone “sutures.” With a higher coefficient of restitution, the kinetic energy of the infant head before impact is transferred back to the head after impact, which sets it in motion in the opposite direction. The lower coefficient of restitution of the toddler head causes a more inelastic collision with the impact surface, minimizing rebound, which may account for the smaller peak-to-peak changes in angular velocity. Figure 8 shows a representative sagittal velocity trace of the toddler impact onto concrete from 1 ft. The peak-to-peak change in angular velocity is calculated from the difference between consecutive maximum (Peak 1) and minimum (Peak 2) velocity peaks (Fig. 8). When there is no rebound, Peak 2 is most often located at or near $\omega = 0$. When rebound occurs, the peak-to-peak change will be greater because of the change in head direction, producing larger negative velocities.

In addition to differences in peak-to-peak change in angular velocity, the absence of head rebound in the toddler may also account for shorter time duration. When the infant head contacts the impact surface, the skull deforms to attenuate the impact, causing the duration of the event to be longer. By comparison, when the toddler head contacts the impact surface, the fused skull deforms very little, causing the impact event to be rapid. The combination of a larger head mass and fused skull may account for the observed differences.

Previous studies of head impact in the toddler have been performed using the Hybrid II 3-year-old dummy. Bertocci et al.⁸ simulated falls from beds and couches by placing the dummy in a supine position and pushing the dummy off the edge of a surface 0.68-m high onto wood, padded carpet, linoleum and playground foam. Linear head acceleration, pelvis acceleration, and femur loading were measured, and HIC values were calculated from linear acceleration. Bertocci et al. concluded that all scenarios produced HIC values below injury thresholds. The range of linear accelerations reported in the Hybrid II 3-year-old dummy (approximately 1000–2500 m/second) is within the range of linear accelerations estimated in our study (638–5173 m/second). In a follow-up study of the effect of wet versus dry linoleum on head injury risk, the same group reported linear head accelerations of approximately 700–1500 m/second from feet-first falls, which also overlaps in the lower range reported in our study.¹⁶ However,

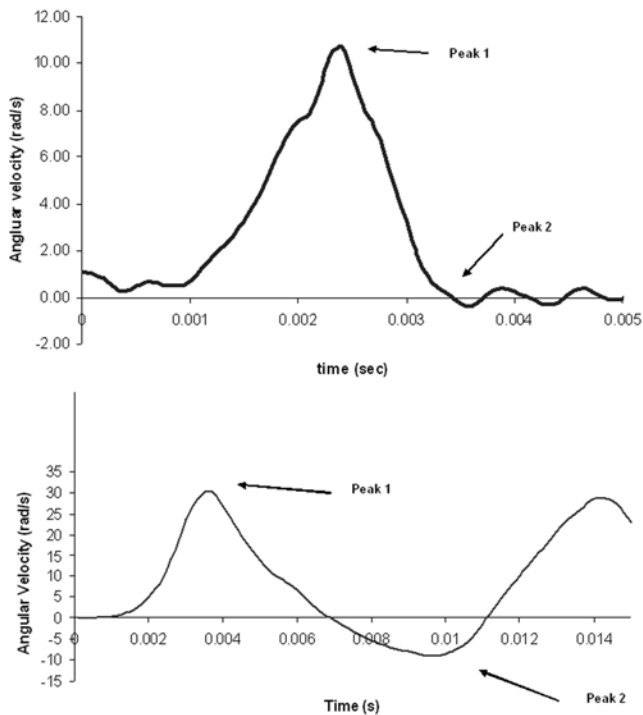


Fig. 8. Angular velocity traces in the sagittal direction from toddler (upper) and infant (lower) drops. Angular acceleration in the toddler consisted of a single peak, whereas in the infant a cyclic pattern was observed due to repetitive head rebound.

the range of linear accelerations in our study of occipital falls extends beyond those reported in feet-first falls, not surprisingly suggesting that head-first falls result in more severe head injuries compared with feet-first falls because the initial impact was with the feet, absorbing some of the impact energy before head contact.

Current National Highway Traffic and Safety Administration standards use the HIC, which is based on linear head acceleration, to develop thresholds for head injury in the pediatric population. The HIC tolerance levels have been scaled from the adult ($HIC_{36} \leq 1000$, $HIC_{15} \leq 700$) to develop thresholds for 12-month-old ($HIC_{36} \leq 660$, $HIC_{15} \leq 390$) and 3-year-old ($HIC_{36} \leq 900$, $HIC_{15} \leq 570$) children based on material properties specific to the cranial sutures but still do not account for the angular acceleration of the head during an event.³³ The resultant linear accelerations from our study yield HIC_{36} values that range from 4.7 (1 ft onto carpet pad) to 42.4 (3 ft onto concrete). However, the time durations in our drops ranged from 2 msec (1 ft onto concrete) to 7 msec (1 ft onto carpet pad), which are significantly shorter than the 15- or 36-msec durations used for HIC calculations. Tolerance levels for such short durations are likely to be even lower than HIC_{15} values. While the HIC values for our data are well below the HIC_{15} head injury thresholds for 12-month-old and 3-year-old children, new HIC thresholds such as HIC_5 are needed to make more reasonable predictions of injury. Moreover, investigators have noted that traumatic head injury is more closely associated with rotational effects (angular acceleration or velocity) rather than translational motion (linear acceleration) of the head.^{50,52,53,57} For this

reason, we chose to measure angular acceleration of the head in the current study. Unfortunately, at present, there are no studies in living or cadaveric toddlers that have investigated head impact and the associated potential injuries from angular head accelerations by which to extend our results to predict head injuries.

There are, however, published studies in adult volunteers that provide a range of head accelerations associated with the presence or absence of loss of consciousness. Pincemille et al.⁵⁹ mounted accelerometers on the helmets of volunteer boxers during 5 training fights and measured accelerations and velocities of a set of 44 blows. All blows were reported to be nonconcussive, and we used these as negative controls. When accelerations from the boxers are mass scaled⁵⁴ according to the inverse ratio of the masses to the two-thirds power and velocities were scaled according to the inverse ratio of the masses to the one-third power from the adult brain (mass = 1440 g)¹⁷ to the toddler brain (mass = 1018 g),¹⁷ we note that toddler drops from 1 and 2 ft onto carpet pad (Fig. 9) are within the range of angular accelerations of the boxer blows that did not result in concussion (Fig. 9), but in terms of angular velocity are, on average, 3 times lower than the boxer blows. The toddler concrete drops, however, are 4 times larger in angular acceleration compared with the measured boxer blows, suggesting the potential for loss of consciousness in toddler low-height, head-first impacts.

For positive control data for concussion, we used data obtained by Pellman et al.⁵⁷ in which digital video was used to extract impact position and velocity of football players who received a concussion while colliding with other players. These collisions were reenacted using commercial adult surrogates and measured the angular accelerations for a series of 15 impacts (Fig. 9). Similar to the comparison with the boxer data, the toddler drops from 1 ft onto carpet pad (Fig. 9) are within the range of angular acceleration of the football impacts, and the toddler drops onto concrete are well above the concussive range. We hypothesize that accelerations from drops in the toddler are at or above the angular acceleration level of boxer blows and football hits due to the restrictive head motion in the drops. For impact events in these athletic settings, it is likely that the head is allowed to follow through with motion after impact and decrease the potential for very high head deceleration. In contrast, the toddler drops represent a worst-case scenario in which the head is forced to come to a rapid stop against the impact surface.

Despite the reports of concussion in the Pellman data, we note significant overlap between the nonconcussion boxer load data and the National Football League load data. Although the literature is not clear on whether velocity or acceleration is more predictive of brain injury, we used the scaled boxer and football data to estimate the predictive capabilities of rotational acceleration and velocity for concussion. The boxer and football data were assigned a binary outcome (0 = no concussion, 1 = concussion), and the area under the receiver operator characteristic curve was used to evaluate the sensitivity and specificity of both rotational acceleration and velocity separately.¹⁹ An area of 0.5 indicates a random response to concussion, whereas higher areas (up to a maximum

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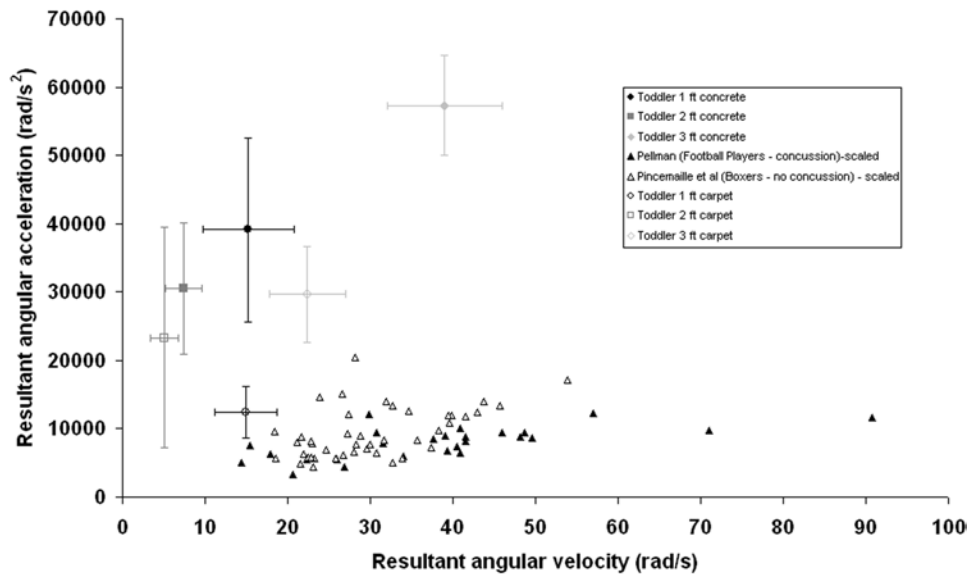


Fig. 9. Resultant acceleration/velocity data from toddler drops compared with mass scaled concussion data from boxers and football players.

of 1) indicate better predictive capability. With receiver operator characteristic curve areas of 0.56 and 0.68 for acceleration and velocity, respectively, velocity is a better predictor of concussion.

We used the scaled velocities from the boxer and football player data to predict concussion in our toddler drops from 1, 2, and 3 ft onto carpet and concrete. The scaled data suggest that there is a 10% chance of concussion if velocity is greater than 14 radians/second and a 20% chance of concussion if the velocity is greater than 24 radians/second. Because the velocities for 3-ft drops onto concrete (39 radians/second) are above this 20% occurrence value, we conclude that concussion is possible from 3-ft falls onto concrete with primary, direct contact to the occiput. Importantly, this analysis assumes that adult data can be scaled to the toddler brain using only brain mass, and that mechanical properties of brain tissue and critical deformations associated with injury are the same across age. Previously, we compared properties of “toddler” and adult porcine brain tissue and determined that they were statistically indistinguishable.⁶⁴ However, if the toddler brain tissue is determined to be more vulnerable to injury (critical deformations are lower) than the adult brain tissue, then the chance of concussion would be higher than estimated above.

Although there are no published concussion tolerance data for very young children, we report the incidence of neurological impairment in a set of 285 infants and toddlers admitted to the hospital for head trauma related to accidental falls.²⁸ Of the 285 patients, 31 were toddlers (1–4 years) who fell from 3 ft or less. In this population, 64.5% of toddlers showed some evidence of altered mental status such as lethargy, sluggishness, unexplained irritability, or loss of consciousness. However, only 31% were noted to have had a concussion (loss of consciousness). The impact surface was noted in 18 cases, of which only 1 was considered a padded surface such as carpet. The majority of falls were onto concrete or ceramic tile.

It may be difficult to make a direct comparison between these clinical data and the drop tests in Fig. 9 because we cannot be certain that direct head impact with a surface occurred in the 31 low-height falls in toddlers. Also, if head impact did occur to cause neurological impairment, it may have been due to the torso or appendage impact or even repeated head impacts in a single fall. However, if we consider only the subset of toddler cases (14) in which evidence of head impact was found (either soft-tissue injury to the head or face or skull fracture), 8 (57.1%) of 14 toddlers experienced altered mental status, and 3 (21.4%) of those 8 had a loss of consciousness. A larger sample size is needed to determine if these frequencies are statistically relevant. These frequencies are higher than anticipated from the adult data in Fig. 9, indicating the potential contributions of lower tissue deformation thresholds in the toddler compared with the adult.

We also report an average impact force associated with each drop height and surface scenario. The overall range for all drops was 2.0–9.5 kN. Although there are no published data for the force required to achieve skull fracture in toddlers, dynamic impact tests in human adult cadavers have reported an average skull fracture force of 11.9 ± 0.9 kN⁷⁷ and a facial fracture force of 2–4 kN.⁴⁸ Coats¹² reported a 50% probability fracture force of 0.28 kN in infants. Because the toddler skull has material properties similar to the infant ($E = 321$ MPa) but is 1.67 times thicker than the infant skull, we can estimate that the structural rigidity of the toddler skull is 1.67³, or 4.6 times that of the infant skull. Therefore, the toddler skull should withstand 4.6 times more force than the infant before fracture. Using data from Coats,¹² we estimate a 50% probability of fracture at 1.288 kN, and based on this estimate, we predict $\geq 50\%$ probability of fracture in all occiput-contact drops from 3 ft or less. However, this incidence of skull fracture from falls ≤ 3 ft represents an extreme head-first contact. Given that limb and torso contact occur frequently and would reduce head impact force,

we expect actual rates of skull fracture from low-height falls to be $\leq 50\%$. Our predictions are corroborated by the clinical cohort of 31 toddlers (mentioned above) who fell from low heights in which nearly 25% had a skull fracture. As expected, the incidence in the clinical data set is lower because some children may not have experienced direct, initial head impact, but broke their fall with limb or torso impact.

Conclusions

Drop tests with an 18-month-old anthropomorphic surrogate show that peak rotational head accelerations in the sagittal direction increase with increasing surface stiffness regardless of height, but that peak rotational acceleration was only height dependent for drops onto carpet pad. The majority of head motion following impact occurred in the sagittal and horizontal directions, with minimal rotation in the coronal plane. The measured accelerations in the toddler surrogate lie in and above the range of previously measured impacts in adult boxers and football players scaled to toddlers, and suggest that falls resulting in direct occipital impact from 3 ft onto concrete may cause concussion.

When comparing the toddler head response to the infant, we observed larger angular acceleration and estimated peak impact force but smaller peak-to-peak change in angular velocity and impact duration in the toddler. We attribute these differences to the larger head and torso mass, stiffer neck, and thicker fused skull. Because of the larger accelerations in the toddler, we expect a higher incidence of neurological impairment in the toddler compared with the infant in direct head impact scenarios. Although the toddler skull is thicker and can withstand greater forces before fracture, the calculated impact force of falls from 1, 2, and 3 ft onto carpet pad and concrete in the toddler are well above the published fracture force for infants and the estimated fracture force for toddlers, indicating that skull fracture can occur in these events.

Taken together, these findings are noteworthy because they demonstrate that the infant and toddler heads experience different mechanical loading during an accidental fall with head impact. If we assume similar fracture limits and brain tissue injury thresholds between infants and toddlers, these differences contribute to age-dependent head injury responses to the same fall event. This work may aid in identifying injury etiology and the design of safety and playground equipment for the prevention of head injury in infants and toddlers.

Disclosure

This study was funded by NSF, through The Center for Childhood Injury Prevention and Safety, as well as the National Institute of Neurological Disorders and Stroke (NIH Grant No. R01-NS-39679; authors receiving funding: Susan S. Margulies and Nicole G. Ibrahim; NIH Grant No. T32-NS-043126; author receiving funding: Nicole G. Ibrahim). Funding provided for salary (Susan S. Margulies) and stipend and tuition (Nicole G. Ibrahim), and research materials.

The authors report no conflict of interest concerning the materials or methods used in this study or the findings specified in this paper.

Author contributions to the study and manuscript preparation include the following. Conception and design: both authors. Acquisition of data: Ibrahim. Analysis and interpretation of data: both authors. Drafting the article: Ibrahim. Critically revising the article: both authors. Reviewed final version of the manuscript and approved it for submission: both authors. Statistical analysis: Ibrahim. Administrative/technical/material support: Margulies. Study supervision: Margulies.

Acknowledgments

The authors acknowledge John Ibrahim and Brittany Coats for their technical assistance.

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Manuscript submitted August 12, 2009.

Accepted March 17, 2010.

Portions of this work were presented in poster form at the 26th Annual National Neurotrauma Symposium, Orlando, Florida, July 27–30, 2008.

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