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Potential for head injuries in infants from low-height falls

Laboratory investigation

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Object. Falls are the most common accident scenario in young children as well as the most common history provided in child abuse cases. Understanding the biomechanics of falls provides clinicians with objective data to aid in their diagnosis of accidental or inflicted trauma. The objective of this study was to determine impact forces and angular accelerations associated with low-height falls in infants.

Methods. An instrumented anthropomorphic infant surrogate was created to measure the forces and 3D angular accelerations associated with falls from low heights (0.3–0.9 m) onto a mattress, carpet pad, or concrete.

Results. Although height significantly increased peak angular acceleration (α_p), change in peak-to-peak angular velocity, time duration associated with the change in velocity, and peak impact force (F_p) for head-first drops onto a carpet pad or concrete, none of these variables were significantly affected by height when dropped onto a mattress. The α_p was not significantly different for drops onto a carpet pad and concrete from 0.6 or 0.9 m due to compression of the carpet pad. Surprisingly, sagittal α_p was equaled or surpassed by axial α_p.

Conclusions. These are the first 3D angular acceleration and impact force data available for head impact in infants from low-height falls. A future study involving a computational model of the infant head will use the loads measured in this study to predict the probability of occipital skull fracture on impact from head-first low-height falls. Together, these studies will provide data that will aid clinicians in the evaluation of accidental and inflicted head injuries, and will contribute to the design of safer environments for children. *(DOI: 10.3171/PED.2008.2.11.321)*

Key Words • **anthropomorphic surrogate** • **biomechanical study** • **child abuse** • **head impact** • **low-height fall** • **pediatric head injury**

Falls are the most common unintentional injury modality among infants 0–3 months old (CDC Wisqar Database, 2005 [http://www.cdc.gov/ncipc/wisqars/]) as well as the most common history provided by caretakers in cases of suspected child abuse.[22] Distinguishing between unintentional and inflicted trauma can be challenging for clinicians because of the disparity in the literature as to what types of injuries can occur in children from low-height accidental falls. Helfer et al.[13] investigated 85 children (< 5 years old) who fell from hospital beds and found only 2 incidences of skull fracture. However, it is unknown how many infants (< 1 year old) were included in this study, what the primary impact site was for these cases, or whether intravenous tubes, blankets, and other hospital peripherals were involved during the falls. Williams[28] reported 3 fractures in children < 3 years old involved in witnessed low-height falls (< 1.5 m) but did not delineate young infants from this larger age group or reveal how many total patients were in the low-height fall group. Regardless, both studies conclude that fracture may occur, but it is rare. In contrast, Tarantino et al.[24] examined 167 infants (< 10 months old) who had a history of falling < 1.2 m. Skull or long bone fracture was found in 11.4% of them, but the study does not eliminate cases of possible abuse. Likewise, Hall et al.[12] reported discovery of skull fractures at autopsy for 100% of children (average age 2.4 years) falling ≤ 0.9 m; however, it was unclear if cases suspicious for child abuse were eliminated.

To evaluate the likelihood of skull fractures from low-height falls, an integrated approach was developed that incorporates measured fall loading conditions with the aid of an anthropomorphic infant surrogate, skull properties and injury threshold data, and the development of an anatomically accurate computational model of the infant head. The combination of these 3 components (Fig. 1) can be validated using well-witnessed real-world accidents. Once validated, the likelihood of skull fracture can be assessed in low-height fall scenarios. Skull response and threshold data as well as the development of a human infant head computational model have been previously published.[4,5]

The focus of the present work was to measure a range

Abbreviations used in this paper: α_p = peak angular acceleration; ANOVA = analysis of variance; DAI = diffuse axonal injury; Δt = time duration associated with change in angular velocity; $\Delta\omega$ = peak-to-peak change in angular velocity; F_p = impact force; ROM = range of motion; SNK = Student-Newman-Keuls.

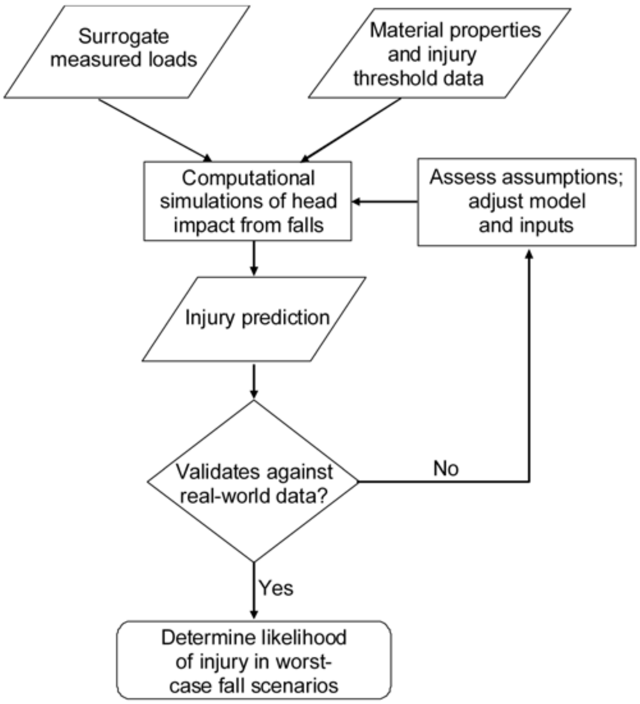

FIG. 1. Flow chart showing how the measured surrogate loads from this study and previously published skull tissue response and threshold data are input in a computational model of the infant head to predict the likelihood of injury in low-height falls.

of angular accelerations and impact forces that occur under worst-case fall scenarios in infants (that is, primary impact of the ground to the occiput of the head without hindrance from extremities). Previously, we performed similar biomechanical studies with child surrogates to investigate biomechanics of head injury in infants. Specifically, Duhaime et al.[7] and Prange et al.[19] developed 4- and 6-week-old infant anthropomorphic surrogates, respectively, to compare the velocity and acceleration responses of shaking to intentional impact onto various surfaces. The surrogate used by Prange and colleagues was designed to measure the maximum rotational acceleration response of the head during these events. The skull case did not include the sutures or separate cranial plates, but rather simulated the infant skull as a single piece of cranial bone. Furthermore, the neck of the surrogate was a resistance-free hinge that limited head motion to only sagittal rotation. Finally, although head mass was represented accurately, the surrogate did not contain extremities and therefore all the remaining body mass was located in the trunk region. Together, these idealizations enabled the authors of those studies to determine worst-case head angular accelerations in low-height falls. However, no measurements were made regarding the impact forces involved in low-height falls.

The current study expands on the aforementioned surrogate models to develop a 1.5-month-old anthropomorphic infant surrogate with realistic skull case, neck, and weight distribution. The 3D angular acceleration and velocity (that is, measurements of sagittal, coronal, and axial rotation of the head), and impact forces were measured to evaluate the biomechanics of head-first low-height falls of young infants. We hypothesized that the addition of sutures to the skull case and the incorporation of a 3D mobile neck would significantly decrease the peak head angular acceleration compared with the surrogate data of Prange et al.[19] with a hinge neck. We also hypothesized that falls from higher heights onto harder surfaces would generate significantly more impact force than falls from lower heights onto softer surfaces.

The 3D angular acceleration, velocity, and impact force data will provide valuable information regarding the head biomechanics of infants involved in low-height falls. High angular acceleration and velocity of the head are often correlated with intracranial hemorrhages and traumatic white matter injury in the brain.[8–11,14,15,21] Large impact forces to the head can cause skull fracture, epidural hemorrhages, and focal contusion to the brain and scalp. Measuring the magnitude of angular acceleration and impact force occurring in low-height falls can lead to a better understanding of possible types of injuries occurring in these scenarios and provide valuable data for the design of safer environments for children.

The surrogate developed in this study will be used in future studies to reenact actual witnessed real-world scenarios. In these future studies the measured loads from these reenactments, in conjunction with the aforementioned skull threshold data and computational model will be used to predict the presence of skull fracture. Good correlation of findings reported in the medical records to the predictions of skull fracture in the model will validate the approach and its components (surrogate construction, tissue thresholds, and computational model design). Once validated, the impact force data measured in this study, combined with the tissue threshold data and the computational model, will be used to predict the likelihood of fracture in worst-case low-height fall scenarios—those resulting in primary occiput contact to the ground with no impedance of the fall by limbs or other objects.

Methods

Anthropomorphic Surrogate Weight and Dimensions

An infant anthropomorphic surrogate (Fig. 2A) was designed to simulate the head response of a 1.5-month-old human infant to low-height falls. Anatomical measurements reported in the literature were used to determine surrogate dimensions and weight distribution of the head, trunk, and extremities (Table 1).

Head and Skull Case

The surrogate skull case was composed of 5 copolymer polypropylene (Boston Brace International Inc.) plates (Fig. 2B) attached together with silicone rubber (Smooth Sil 950, Smooth-On). The stiffness of copolymer polypropylene and silicone rubber was not significantly different from that measured in human infant parietal bone and coronal suture, respectively.[3] The completed skull case was attached to the facial and basal portion of a You & Me doll (KS Toys, Ltd.) with brass screws (Fig. 2B). A 1-mm thick-latex cap, previously shown to be an adequate model of scalp,[18] was placed over the skull case. The periphery of

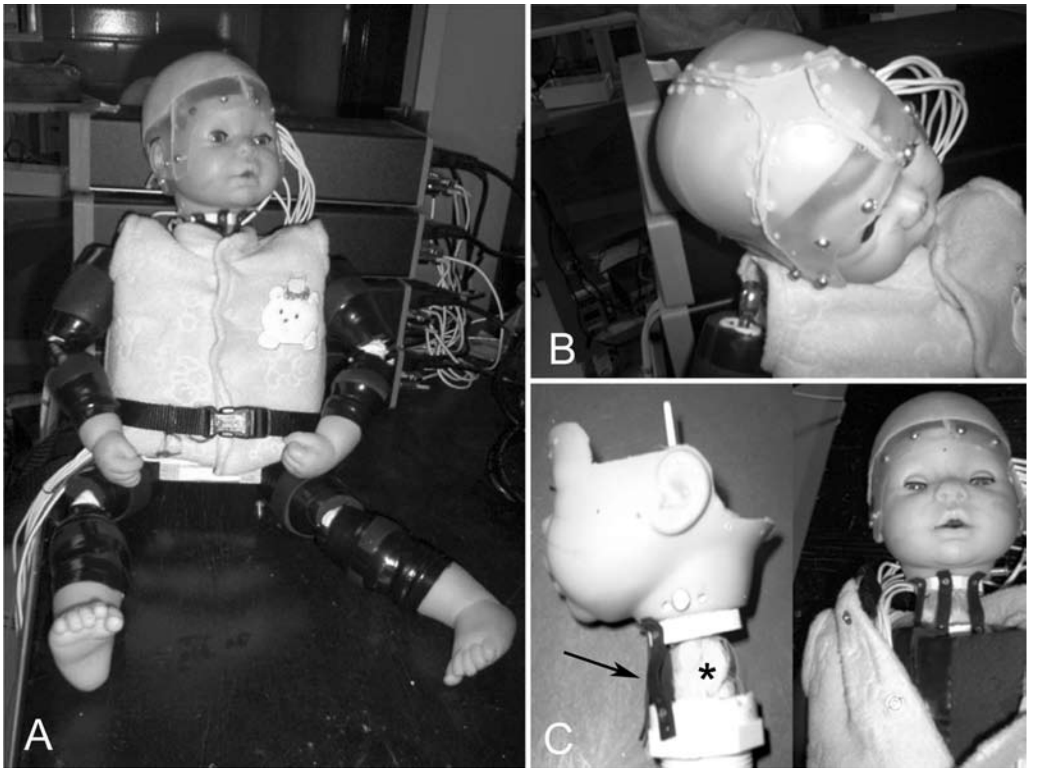

FIG. 2. Photographs showing the anthropomorphic infant surrogate designed to measure head angular acceleration, angular velocity, and impact force resulting from low-height falls (A). The surrogate has copolymer plates and silicone rubber with a stiffness similar to infant human skull and suture and has (C) nylon rope *(asterisk)* for 3D mobility and frontal rubber strips *(arrow)* to add stiffness to the neck in extension.

the latex cap was secured to the head of the infant model with electrical tape.

An accelerometer mount was rigidly attached to a metal plate located in the center of the surrogate's head (Fig. 3). The mount was composed of 9 linear accelerometers (7264B-2000, Endevco), 3 in each x-y-z direction. An angular velocity transducer (ATA Sensors) was also attached to the mount to measure sagittal rotation.

Neck

There is a paucity of data regarding the tension, compression, and bending material properties of the infant pediatric neck. However, Prange and Myers recently measured the bending moment of the C4–5 segment from an unembalmed 24-day-old infant cadaveric osteoligamentous cervical spine (unpublished data, 2004). The nondestructive test protocol investigated quasistatic bending of the spine motion segment from ~ −0.14 Nm (extension) to 0.07 Nm (flexion) (*diamonds* in Fig. 4).

To simulate these properties, the neck of the surrogate was created to be flexible in 3 directions with no fixed center of rotation. The ends of a 1.9-cm-diameter nylon rope were embedded in two 3.8-cm-diameter plastic pipe fittings using Plaster of Paris, creating a flexible 2.9-cm-long neck (Fig. 2C). The entire neck structure (from one pipe fitting to the other) was tested in flexion and extension and compared with the unpublished human infant data. Three 4-cm-long strips of rubber (1 mm thick) were attached to the front of the neck to add stiffening in extension to approximate the results of Prange and Myers data (provided via personal communication, 2004). The overall stiffness of the surrogate neck (*circles* in Fig. 4) mimicked the quasistatic data of Prange and Myers for a C4–5 segment undergoing anteroposterior flexion and extension. It is not known how a child's neck responds in lateral flexion/extension or axial torsion. Therefore, the surrogate neck was created to have minimal resistance to motion in these directions, which may overestimate coronal and axial rotational responses.

The rope by itself exhibits some stiffness (*open circles* in Fig. 4), but it is small and easily overwhelmed by the head weight. Thus, the surrogate neck does not support the weight of the head and it readily flops over when the surrogate is positioned upright. The 3D ROM of the surrogate head in sagittal, axial, and coronal rotation is 132° (90° flexion, 42° extension), 103°, and 73°, respectively. A recent study measuring passive head ROM in 38 infants (2–10 months old) reported similar values for the average ROM for a 2-month-old infant (105° for axial rotation, 68° for coronal rotation).[16] Sagittal ROM was not reported in that study.

Torso and Extremities

The extremities of the surrogate were composed of hollow metal rods weighted with lead balls and covered with cotton cloth and 1.3-cm-thick polyethylene insulation foam. Ball-and-socket joints attached all extremities to the wood torso frame of the surrogate. Ninety-degree motion

TABLE 1

*Dimensions and weight distribution of the anthropomorphic surrogate**

Variable	Human Infant†	Surrogate
dimensions (cm)		
newborn head circumference	33.0–35.6	34.3
infant head breadth (rt/lt)	10.4	10.2
infant head length (anterior/posterior)	13.9	10.2
infant top of head to mid ear	9.6	8.7
infant mid ear to top of neck	5.4	4.3
infant top of head to chin	13.6	12.2
neck length (mobile portion only)	NA	2.9
infant shoulder breadth	17.6	17.8
infant chest depth	9.5	8.2
torso length	25.1	21.9
upper-extremity length (shoulder-wrist)	25.8	27.3
lower-extremity length (hip-heel)	28.8	29.8
weight distribution		
total body weight (kg)	4.8	4.4
head weight (kg)	0.77–0.87	1.0
head/body ratio	0.23	0.23
upper extremity/body weight ratio	0.08	0.09
lower extremity/body weight ratio	0.15	0.14

* NA = not applicable.

† The head circumference was based on measurements of a newborn according to Chadwick, et al. The torso length was calculated from a 0–3-month-old infant according to Ohman and Beckung, and the extremity lengths were based on measurements of a 0–2-month-old infant according to Pellmen et al. All other dimensions were measured in a 0–3-month-old infant according to Ohman and Beckung. The weight distribution was measured in a 1-month-old infant according to Cory and Jones, except for the extremity/weight ratios, which were measured in a newborn.

pin joints connected the lower and upper regions of the extremities to simulate bending of elbow and knee joints. A 3-accelerometer array was placed in the surrogate torso frame to measure the 3D linear acceleration of the torso. Polyethylene foam was used around the torso to protect the instrumentation and frame from impact damage.

Drop Testing Protocol

The instrumented anthropomorphic surrogate infant underwent a series of head-first drop tests from 3 heights (0.3, 0.6, and 0.9 m) onto 3 surfaces (concrete, 0.6-cm-thick carpet pad, and 15-cm-thick innerspring crib mattress covered with a cotton sheet). The innerspring crib mattress had a thin plastic outer covering, followed by 2.5-cm layer of soft foam. The inner structure of the mattress was spanned by 7.6-cm-long springs. The average linear elastic moduli for a 40% deformation of the carpet pad and intact innerspring crib mattress were measured as 621 and 0.690 kPa, respectively. The surrogate was suspended in a supine position with the legs outstretched and arms secured to the sides of the body (Fig. 5). Simulating a Moro reflex in the surrogate (arms outstretched) may decrease the measured accelerations and impact forces slightly, but this configuration was selected to simulate a worst-case scenario with initial contact of the ground made to the occiput of the head without hindrance of the extremities.

The surrogate was dropped 10 times for each possible combination of height and floor surface resulting in a total of 90 drops. Digital video (29.97 fps, DCR-TRV103, Sony USA) was used to observe the kinematic response of the surrogate to impact.

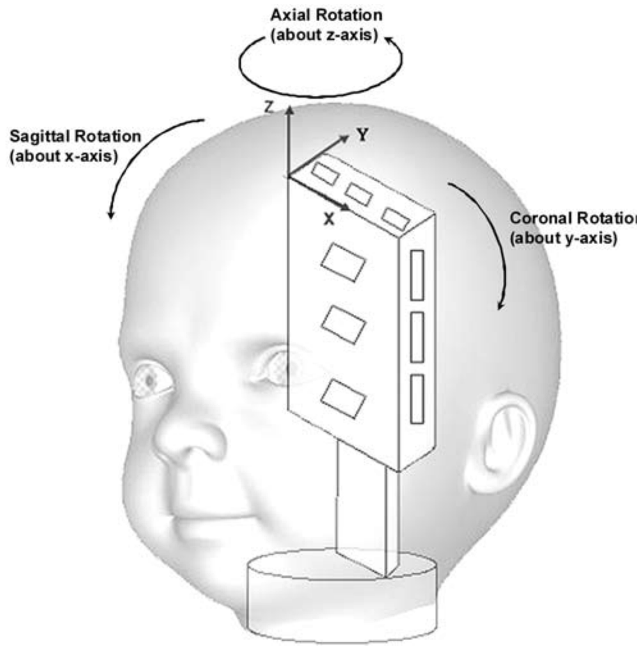

FIG. 3. Schematic of the 9-accelerometer array within the surrogate's head. Rotation around the x, y, and z axes was defined as sagittal, coronal, and axial motion, respectively.

Data Analysis

All 13 sensor outputs (1 angular velocity and 12 linear accelerations) were collected using a computer data acquisition system (Labview 4.1, National Instruments and Dell) at 10,000 samples per second. All acceleration and velocity data were filtered according to SAE J211–1 standard for automotive crashes.[23] However, falls onto mattresses generated smaller frequencies than the standard 1640-Hz cutoff, so a spectral analysis was used to determine a more appropriate cutoff frequency of 300 Hz. Peak acceleration for the adjusted cutoff frequency was compared before and after filtering to verify that no important peak data were lost in filtering.

To determine the 3D angular velocity and acceleration of the surrogate's head, a program was created in MATLAB (MathWorks) that optimized rigid body equations relating to linear and angular acceleration and velocity. A similar program (developed in Excel Visual Basic) has been previously validated and used (E. Takounts et al., unpublished data presented at the Injury Biomechanics Research, 31st International Workshop, 2003). The angular velocity measured in the head was used to validate the calculated angular acceleration in sagittal rotation (rotation about the x axis, α_x).

Nine inertial output measurements were analyzed: maximum peak angular acceleration (α_{px}, α_{py}, α_{pz}), maximum peak-to-peak change in angular velocity ($\Delta\varpi_x$, $\Delta\varpi_y$, $\Delta\varpi_z$), and the maximum time interval for peak-to-peak angular velocity (Δt_x, Δt_y, Δt_z) in the sagittal (rotation about x), coronal (rotation about y), and axial (rotation about z) directions. Separate 2-way ANOVAs, with height and surface as the factors affecting these output measurements, were used to determine significant differences among the drop height-surface combinations.

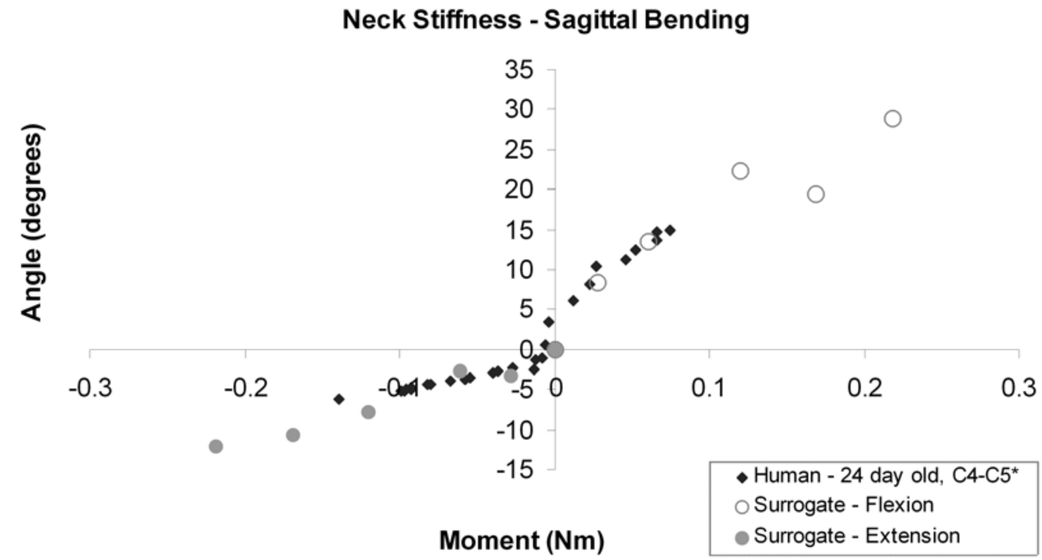

FIG. 4. Scatterplot showing the moment measurements in flexion and extension of an unembalmed infant C4–5 segment compared with the surrogate. The *asterisk* indicates unpublished human data provided via personal communication from B. Myers and M. Prange, 2004.

For each drop height-surface combination, impact force was calculated using Newton's second law: F = ma with a = rα, where "m" is the mass of the surrogate head, "a" is the linear acceleration of the surrogate head upon impact, "r" is the distance from the center of mass of the head to the point of rotation, and "α" is the largest α_p across all directions.

Because the flexible neck of the surrogate had no single point of rotation, impact forces calculated using the bottom and top of the surrogate neck represent the upper and lower limits of the force range, respectively. The peak F_p of the upper limit values (representing the worst-case scenario) for each drop height-surface combination was used in a 2-way ANOVA to determine the significant effects of height and surface on impact force. An SNK test was used to perform multiple comparisons among all the groups for all the output measurements. A Type I error of 5% was used to determine significance for all tests.

Results

With rotation around the x, y, and z corresponding to sagittal, coronal, and axial rotations, nine inertial output measurements (α_{px}, α_{py}, α_{pz}, $\Delta\varpi_x$, $\Delta\varpi_y$, $\Delta\varpi_z$, Δt_x, Δt_y, Δt_z, and peak impact forces (F_p) were determined from the acceleration and velocity data for a total of 90 drops. A typical fall consisted of an initial impact to the occiput of the head, followed by an impact of the torso and legs as the head began to rotate forward. The head and torso acceleration data and video images confirmed that the midline of the occiput consistently made contact with the ground before the torso on all surfaces. All data analysis focused on the initial impact of the head.

Initial contact produced head deceleration as the head impacted the ground, followed by a rapid acceleration as the head rebounded forward. The peak acceleration value was occasionally larger than the initial deceleration value, most often in falls from 0.6–0.9 m onto carpet and concrete. The peak angular acceleration for each direction (α_{px}, α_{py}, α_{pz}) used in analysis was chosen as whichever was the larger value of the peak acceleration or deceleration associated with these initial contact events.

A 2-factorial ANOVA was used to determine significant effects from 2 factors (surface and height) and an SNK test was used to perform multiple comparisons among all the groups for the output measurement that were normally distributed: angular velocity ($\Delta\varpi_x$, $\Delta\varpi_y$, and $\Delta\varpi_z$) and F_p. The remaining variables (α_{px}, α_{py}, α_{pz}, Δt_x, Δt_y, and Δt_z) were not normally distributed and nonparametric equivalents of the ANOVA and SNK test were used. The nonparametric tests involved ranking all 90 data points in each output measurement and performing standard ANOVA procedures to the ranks. For multiple comparisons, the ranks in each surface-height combination were summed, and SNK procedures were applied to these sums.[30]

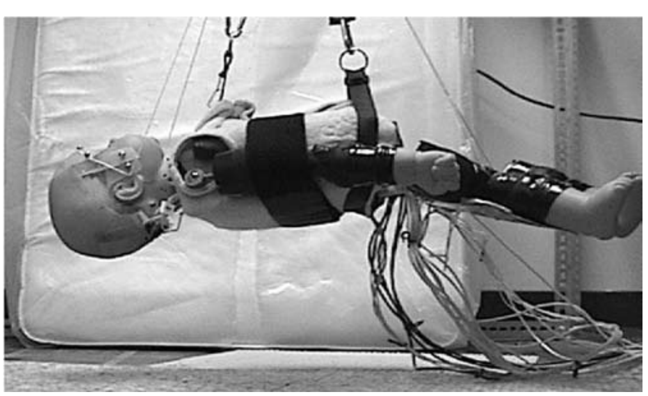

FIG. 5. Photograph representing a worst-case scenario. The surrogate was suspended and released in a supine position with legs extended and arms fixed to the torso. This configuration resulted in initial ground contact to occur with the head of the surrogate.

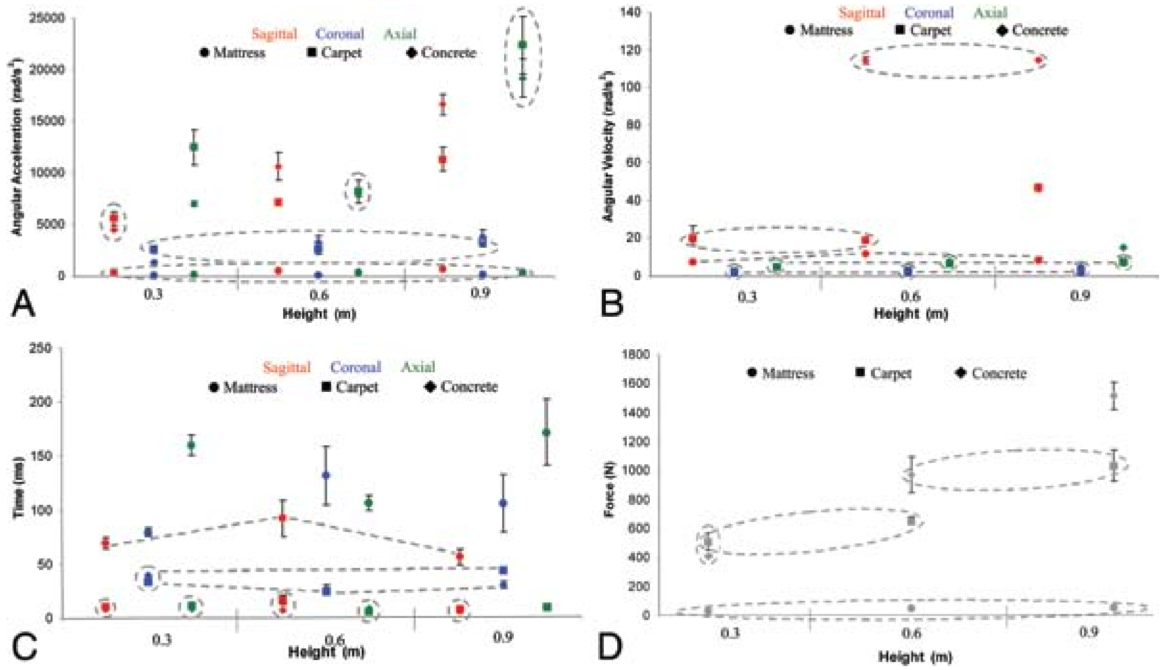

FIG. 6. Scatterplots showing the mean ± standard error of the mean of angular accelerations (A), peak-to-peak change in angular velocity (B), time duration during peak-to-peak change in angular velocity (C), and peak force (D) from all heights (0.3–0.9 m) onto all surfaces (mattress, carpet, and concrete) measured in all rotational directions (sagittal, coronal, and axial). *Dotted lines* indicate groups that were not significantly different. *Whiskers* indicate the standard error of the mean.

Peak Angular Acceleration

Peak angular acceleration or deceleration (α_p) was at least 2 times larger in sagittal and axial rotation than in coronal rotation for all surfaces and heights (Fig. 6A). A 2-factorial (height and surface) ANOVA found an increase in height to significantly increase α_p in the sagittal and coronal directions ($p < 0.001$). No significant difference was found between axial α_p from 0.3- and 0.6-m drops, but drops from 0.9 m had significantly lower axial α_p ($p < 0.001$). Drops onto stiffer surfaces (carpet and concrete) had significantly ($p < 0.001$) larger α_p in all 3 rotational directions (sagittal, coronal, and axial) than drops onto the softer surface (mattress). Mattress was the only material not significantly affected by height in all rotational directions. Additionally, carpet was not significantly affected by height in coronal rotation.

The interaction between height and surface had a significant effect on α_p ($p < 0.001$). For coronal and axial rotation, the α_p of a drop onto a mattress was not significantly different between 0.6 and 0.9 m, but a drop from 0.3 m was significantly lower ($p < 0.05$). Additionally, no significant difference was found between the α_p of drops onto carpet and concrete for any of the 3 directions, except for drops from 0.3 m, which had significantly higher coronal and axial α_p when dropped onto carpet compared with concrete ($p < 0.05$). The increased frictional force between the scalp and the carpet pad created a fixed fulcrum that forced the surrogate head to rotate rapidly about this point rather than allowing the head to slide along the surface during rotation as noticed in drops onto concrete.

Peak-to-Peak Change in Angular Velocity

More than half (53%) of the total measured $\Delta\omega$ occurred in the sagittal direction, and this percentage increased with an increase in impact surface stiffness and height (Fig. 6B). A 2-way ANOVA found sagittal $\Delta\omega$ to increase significantly with an increase in height ($p < 0.01$) and surface stiffness ($p < 0.001$). The interaction between surface and height had a significant effect ($p < 0.001$) with drops onto carpet being significantly larger at 0.9 m than at 0.6 or 0.3 m, and drops onto concrete being significantly lower at 0.3 m than at 0.6 or 0.9 m. Sagittal $\Delta\omega$ experienced from drops onto mattresses were not affected by height at all. Coronal and axial $\Delta\omega$ were not significantly affected by height or surface except for drops from 0.9 m onto concrete, which were significantly higher than all other drops ($p < 0.001$).

Time Duration for Peak-to-Peak Change in Angular Velocity

Longer time durations, Δt, were consistently found in the coronal and axial directions compared with the sagittal direction (Fig. 6C). A 2-way ANOVA found height to be a significant factor of duration with an increase of height and surface stiffness significantly decreasing sagittal Δt ($p < 0.001$).

Coronal and axial Δt measured from drops onto a mattress were significantly longer than drops onto either carpet or concrete ($p < 0.001$), which were not significantly different from each other. Height was a significant factor of Δt in these 2 directions ($p < 0.05$), but no discernable trend could be determined as drops from 0.6 m had a significantly shorter Δt than drops from 0.3 and 0.9 m in both directions.

The interaction between surface and height significantly affected Δt for all rotation directions ($p < 0.01$). Duration during drops onto a mattress was not significantly affected by height in any of the 3 directions. The sagittal and coronal Δt from drops onto carpet and axial Δt from drops onto concrete were also not significantly affected by height.

Peak Impact Force

A 2-way ANOVA found an increase in surface stiffness and drop height to significantly increase the peak F_p ($p < 0.001$, Fig. 6D). The interaction between surface and height significantly affected F_p ($p < 0.001$), and mattress was the only surface that was not significantly affected by height. Peak force from surrogate drops onto concrete were significantly larger than those onto carpet ($p < 0.001$), except from 0.3 m where no significant difference was found between the two. Impact force onto carpet from 0.3 and 0.6 m was not significantly different, but it significantly increased at 0.9 m ($p < 0.001$). Impact force of the surrogate's head onto concrete from 0.6 m was not significantly different than the impact force onto carpet from 0.9 m.

Discussion

Falls are the most common type of accident in young children as well as the most common history provided by caretakers suspected of child abuse. The present surrogate study is the first published 3D head angular acceleration and velocity, and contact force data experienced at head impact in falls of children.[6,7,19,29] As a first principle, larger impact forces have a higher likelihood of producing skull fracture, but tissue thresholds—the critical stresses above which fracture occurs—are the necessary link to associate the measured loads of the surrogate with injury. Thus, the following 2 tools are needed to predict fracture from the measured loads in this study: 1) one that relates fall impact forces to skull stresses (computational models), and 2) one that relates skull stresses to fracture (mechanical properties). Previously, we measured the mechanical properties of skull and suture[5] and constructed a computational model of the pediatric skull,[4] so the 2 tools are in hand. The next step is to use the measured head velocities and forces of this study as input to the computational model and calculate stress everywhere in the skull throughout the time course of the impact event. The calculated stresses are then compared with the mechanical property and failure stresses of skull specimens at autopsy to predict the likelihood of skull fracture. However, before generalizing these predictions, the penultimate step is to reenact specific patient cases with detailed histories and to validate predictions for skull fracture against the radiological reports for each case. Accomplishing this ultimate goal of identifying skull fracture risk in various fall scenarios provides objective data to aid clinicians in their differential diagnoses of children presenting with a history of a fall.

Effect of Neck, Skull Case, and Weight Distribution on Head Acceleration

Previous studies in our laboratory[19] have taken important first steps toward understanding the biomechanics of infants during low-height falls, but they have been limited by a paucity of biomechanical data. This study expands on previous work by incorporating new material property data[5] to develop a more lifelike anthropomorphic surrogate of a 1.5-month-old infant, and thus make more realistic measurements of the head's response to occipital head-first impact following low-height falls.

The peak α_p and $\Delta\omega$ decreased (36–88%) and Δt increased (13–77%) markedly when comparing the values measured with the present surrogate to those reported by Prange et al.[19] There are 3 new features that differ from the Prange surrogate that contribute to these lower values: a more realistic neck, increased deformable skull case incorporating suture, and the inclusion of extremities to create a reasonable body weight distribution. The frictionless hinged neck of the Prange surrogate was designed with negligible neck flexion resistance and restricted rotation to the sagittal direction. This allowed the highest head accelerations possible and identified an upper limit of the infant head acceleration response to impact. The neck used in the present surrogate was designed to permit a lifelike motion that is a combination of sagittal, coronal, and axial head rotation. Importantly, the neck response matched previously unpublished data obtained in human infant cadavers (Fig. 4). Because the surrogate neck had more sagittal resistance than the hinge used in Prange et al. and allowed motion and acceleration to be distributed to other directions, the values of head sagittal α_p and $\Delta\omega$ of the surrogate were significantly below the mean values reported by Prange et al. To determine a single angular acceleration magnitude that accounts for all 3 directions of rotation, a resultant angular acceleration time curve was computed for each drop onto each surface, and the peak value of this resultant curve was identified. The average peak resultant angular acceleration for a 0.9-m drop onto concrete was 22,500 rad/second^2, still 70% lower than that measured by Prange et al. for the same conditions, indicating that the change from a unidirectional neck to a multidirectional neck in these fall simulations is not the sole contributor for the decrease in measured angular acceleration from the Prange surrogate to the present surrogate.

The skull is another contributor to the differences in angular acceleration and velocity between the Prange surrogate and our present model at impact. An infant's skull is composed of bone plates attached by suture and fontanels. The skull case of the current surrogate was designed to mimic the deformable multicomponent skull case of the infant, which produced more deformation of the skull and increased absorption of energy at impact in the surrogate and thus increased the overall duration of the impact event and decreased α_p and $\Delta\omega$ compared with the continuous copolymer skull used by Prange et al.[19]

The addition of extremities and appropriate body weight distribution also played a role in the decreased inertial rotation of the surrogate head after head-first impact. Digital videos from drops of the Prange surrogate (in which no extremities were used) reveal that the distal portion of the torso rotated rapidly toward the head of the surrogate following torso impact onto firm surfaces. Digital videos from drops onto firm surfaces with our surrogate revealed that the legs of the surrogate weighed the distal torso down, causing the body of the surrogate to have very little anteroposterior rotational rebound upon impact, but rather move upward in a translational manner.

Effect of Surface and Height on Head Acceleration

Peak acceleration from occiput-first falls onto an innerspring crib mattress was not affected by the height in any direction, because the compliant pocketing nature of the mattress overwhelmed any influence of fall height. From video, we observed that when the surrogate impacted the mattress, the immediate pocketing of the head persisted long enough for the torso and limbs to contact the mattress. The surrogate head and body then rebounded upward as a unit with minimal angular motion. It is only after a mattress has compressed fully that one would expect the body motion to change. In our study, full compression did not occur for falls $\leq$ 0.9 m. Prange et al.[19] also found no significant effect of height on α_p or $\Delta\omega$ for drops onto a 10-cm uncovered foam mattress. The highest height tested in the Prange study was 1.5 m. Combining these studies, it appears that height has little effect on angular motion when the drop is from 1.5 m or less onto a foam or innerspring crib mattress.

Another overall trend was that at 0.6 and 0.9 m the carpet pad resulted in α_p or $\Delta\omega$ responses that were statistically indistinguishable from concrete. Unlike the 15-cm crib mattress, the carpet pad is only 0.6 cm thick, and we hypothesize that full compression of the carpet pad occurred at both 0.6 and 0.9 m. Thus, at these drop heights the response was wholly influenced by the stiffness of the surface underneath the carpet pad (concrete), and the pad offered minimal protection.

Effect of Head Rotation Direction on Head Acceleration

The surrogate in this study incorporated a novel 9-accelerometer array configuration that allowed 3D motion of the head to be calculated within the small confines of the pediatric head, similar to larger adult commercial surrogates. From this motion analysis, it was found that sagittal rotation was the dominant motion for α_p, $\Delta\omega$, and Δt in these occiput-first falls. Interestingly, however, axial α_p values of the head were as high as, or often slightly higher than sagittal α_p. Axial $\Delta\omega$ and Δt, however, were remarkably lower than those measured in the sagittal direction. Digital video confirmed that upon occipital impact, the surrogate head rapidly rotated in the horizontal plane (axial rotation) but did not subsequently rebound and reverse its rotation. The ovoid dorsal portion of the occiput presents a metastable contact surface, rotating the head axially to the flatter parietal side of the surrogate's head. The frictional force between the surrogate scalp and the impact surface prevents the head from sliding laterally, but rather grips the head at the occiput causing a rapid axial rotation about this pivot point. The friction and loss of energy at this point in the impact prevent the head from rebounding in the reverse direction. Additionally, due to an absence of data in the literature about the infant neck's resistance to axial rotation, the surrogate neck was designed with minimal resistance in this direction, allowing a worst-case scenario to be mimicked. The rapid motion of the head accompanied by minimal axial resistance in the neck explains why there is a high axial α_p with a low $\Delta\omega$. The measured coronal α_p and $\Delta\omega$ were at least 2 times lower than those measured in the sagittal and axial directions, indicating that with frictional surfaces coronal head rotation does not appear to have a large role in falls from heights < 0.9 m.

An increase in height and surface stiffness each significantly increased sagittal α_p and $\Delta\omega$ and decreased sagittal Δt, but these trends were not so clear cut for coronal and axial rotation, which was more affected by the interaction between height and surface. One example is that axial α_p significantly increased with an increase in surface stiffness but not at the higher drop heights. Instead, it was found that there was no significant difference in axial α_p between 0.6- and 0.9-m drops. The axial ROM of the surrogate's neck is limited by the bands of rubber that stiffen during extension of the neck as well as by the finite distance that the head can turn before the surrogate's cheek comes in contact with the surface. Both of these features contribute the angular motion that was likely maximized at 0.6 m and then again at 0.9 m.

Traumatic Brain Injury Risk

It is tempting to try to relate these surrogate responses to injury risk, but no pediatric threshold data exist relating infant head α_p or $\Delta\omega$ to traumatic brain injury. Two head injury studies have investigated the association of concussion in adult professional football players[17] and boxers[1] to angular accelerations following impact similar to the acceleration levels measured in our surrogate drops. The boxer study and the multipart football study reported conflicting results such that no concussion was found in 5 boxers (with instrumented head gear) experiencing levels of head angular acceleration at or above the head angular accelerations (determined from digital video and surrogate reconstructions) associated with concussion in football players. It has been suggested that the discrepancy is due to the different kinematics of the impact experienced in these sports.[25] Even so, the prediction of injury in the pediatric population using these adult data are further confounded by the paucity of information regarding scaling injury thresholds from adults to children. Based on size alone, tolerable levels of acceleration might be expected to be higher in children, but tissue properties[5,20] also play a critical role in scaling adult data to children. We conclude that a more complete comparison between adult and pediatric tissue injury tolerances is required before any adult head injury data can be appropriately scaled to infants.

Gennarelli et al.[9] applied rotational loads to 39 adult primates in either the coronal, sagittal, or axial direction. Scaling issues and the much higher angular acceleration levels in the primate study prevent us from associating the loads measured from the surrogate drops to concussion, acute subdural hematoma, or DAI. However, this primate study provides insight into the importance of rotational direction on injury. To summarize briefly, all 13 animals with coronal rotation experienced lengths of unconsciousness > 6 hours, while all 13 animals with sagittal rotation and 7 of 13 animals with axial rotation had a loss of consciousness for only < 5 minutes. In addition, the majority of the primates undergoing coronal rotation had severe DAI, whereas primates with axial and sagittal rotation had moderate and minimal DAI, respectively. In our surrogate drops from all heights onto all surfaces, coronal α_p was minimal when compared with sagittal and axial α_p. The largest measured coronal α_p was 8389 rad/second^2 from a 0.9-m drop onto concrete compared with 21,640 rad/second^2 in the sagittal direction and 38,860 rad/second^2 in the axial direction. However, because there are no data regarding torsional neck stiffness in the child, the axial rotations may be overestimated by our flexible neck. Nevertheless, axial motion often had larger angular accelerations than the sagittal and coronal, and although this rotational direction was not associated with as severe a brain injury as following coronal rotation, it still produced moderate brain injury in primates. Future pediatric neck studies should focus on torsional stiffness, and future animal brain injury threshold studies should focus on axial rotation because this is one of the dominant motions during occiput-first falls and, according to the primate studies, has the ability to cause longer periods of unconsciousness than sagittal rotation in adults.

Impact Force and Predictions of Skull Fracture

In addition to calculating rotational loads following occiput-first low-height falls, we determined the impact force of the surrogate head on each surface. As might be expected, the peak force significantly increased with an increase in surface stiffness and drop height. However, the interaction between surface stiffness and drop height also had a significant effect on impact force. The head impact force from falls onto a mattress was the only surface not significantly affected by height. Similar to the acceleration response, the mattress allows more deformation on impact and cushions the head of the surrogate, decreasing the differences in impact forces at different drop heights. Furthermore, there was no significant difference between impact force from falls onto carpet at 0.3 and 0.6 m, but the difference was significantly greater at 0.9 m. In contrast to the mattress, the carpet pad is only 0.6-cm thick, and at higher heights the carpet pad was completely compressed and the stiffness of the concrete flooring underneath the carpet pad dominated the impact force response.

Several retrospective case studies exist in the literature that attempt to determine whether fracture can occur at low-height falls.[2,24,28] However, comparison of these studies to the loads measured with the surrogate is confounded by their possible inclusion of abuse cases, large age group ranges, and the inclusion of non–head-first falls. A more direct comparison can be made to Weber's studies involving occiput-first drops of infant cadavers. Weber[27] dropped 15 infant cadavers (age range 0–8.1 months, mean 3.4 months) occiput-first from 0.8 m onto 1 of 3 surfaces: stone/tile, carpeting (0.6 cm), and padded linoleum (1.2 cm). Linear fractures occurred in all 15 cadavers, regardless of surface. In a continuation of his previous work, Weber[26] dropped an additional 35 infant cadavers from the same height (0.8 m) onto 1 of 2 surfaces: a 2-cm foam mat and an 8-cm folded camelhair blanket. Only 2 of the 10 drops onto the foam mat and 4 of 25 drops onto the camelhair blanket resulted in linear fractures. In comparing Weber's series of drops with the present study, our 0.9-m drops of the surrogate onto concrete and carpet pad are the most comparable in drop height and surface conditions. Assuming that the forces calculated in our study are similar to those experienced from the drops in the Weber study, and assuming that lack of head pressurization in the cadavers and their storage and handling have little effect on the occurrence of skull fracture, falls resulting in peak impact forces of 930–1600 N would likely result in a linear skull fracture in infants. From our data in unimpeded occiput-first falls, this would include falls from 0.6–0.9 m onto concrete and falls from 0.9 m onto carpet. Our 0.9-m drops onto a 15-cm crib mattress, which resulted in an average impact force of 56 N, are not similar to any of Weber's conditions, so no comparison can be made from this height onto this surface. We caution the reader that because the assumptions used in the comparison of our surrogate loads to the Weber cadaver data cannot be confirmed, further studies are needed to validate the association of these loads to skull fracture. Therefore, we do not advocate the use of this load range to predict skull fracture in real-world events.

Instead, a conservative determination of likelihood of skull fracture from low-height falls will be produced by combining the upper limits of the loads measured in this study with previously published biomechanical properties for infant skull[5] and a previously published computational model of the pediatric head[4] to predict the probability of skull fracture. Validation of these predictions will be performed with the medical records of well-witnessed accidental low-height falls. Once validated, the upper-limit impact forces of this study will be used in simulations of occipital head-first falls to determine the likelihood of skull fracture in these worst-case scenarios. If absence of skull fracture is predicted in these worst-case scenarios, we would conclude that fracture is unlikely to occur in fall scenarios with lower impact forces (for example lower heights, softer surfaces, impact of the extremities to the ground prior impact of the head, and history of a caretaker inhibiting the fall path).

Conclusions

Overall, an increase in height and an increase in surface stiffness increased sagittal α_p, $\Delta\omega$, and F_p. In comparing the present surrogate study to that of a previous surrogate study from our laboratory,[19] the sagittal α_p and $\Delta\omega$ were much lower with the present surrogate because of the more lifelike features, including a deformable skull/suture skull case, 3D mobile neck, and the addition of extremities with appropriate weight distribution. Additionally, 3D angular acceleration analyses revealed that although coronal rotation was minimal following impact, surprisingly axial head rotation had slightly higher peak angular accelerations than sagittal head rotation, likely due to the natural convexity of the occiput, and may be further enhanced by the laterally laxity of the neck design. There is a paucity of injury data at these low levels of angular acceleration, but previous adult primate studies at much higher accelerations have shown that severity of concussion and DAI are influenced by rotational direction. Future surrogate and animal studies are needed to examine the rotational loads caused by nonocciput impacts from low-height falls and develop direction-specific injury thresholds for the infant at head angular acceleration levels experienced following impact from low-height falls.

Comparison of surrogate drop scenarios in this study to cadaver drop studies[26,27] indicates that linear fracture may occur in the infant from head-first fall heights 0.9 m onto carpet and 0.6–0.9 m onto concrete. However, due to the assumptions made when comparing the measured surrogate loads to cadaver data, the impact force data from this study and our previously published skull tissue response and threshold data will be used in a validated infant head computational model to predict the probability of skull fracture in occipital head-first falls from 0.3–0.9 m onto concrete, carpet pad, and mattress. These injury risk predictions will help clinicians make informed decisions regarding the plausibility of skull fracture with an associated history of a low-height fall, and will provide valuable data for designs of safer environments for children.

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