

Anthropomorphic simulations of falls, shakes, and inflicted impacts in infants

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Object. Rotational loading conditions have been shown to produce subdural hemorrhage and diffuse axonal injury. No experimental data are available with which to compare the rotational response of the head of an infant during accidental and inflicted head injuries. The authors sought to compare rotational deceleration sustained by the head among free falls, from different heights onto different surfaces, with those sustained during shaking and inflicted impact.

Methods. An anthropomorphic surrogate of a 1.5-month-old human infant was constructed and used to simulate falls from 0.3 m (1 ft), 0.9 m (3 ft), and 1.5 m (5 ft), as well as vigorous shaking and inflicted head impact. During falls, the surrogate experienced occipital contact against a concrete surface, carpet pad, or foam mattress. For shakes, investigators repeatedly shook the surrogate in an anteroposterior plane; inflicted impact was defined as the terminal portion of a vigorous shake, in which the surrogate's occiput made contact with a rigid or padded surface. Rotational velocity was recorded directly and the maximum (peak–peak) change in angular velocity ($\Delta\dot{\theta}_{\max}$) and the peak angular acceleration ($\ddot{\theta}_{\max}$) were calculated.

Analysis of variance revealed significant increases in the $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ associated with falls onto harder surfaces and from higher heights. During inflicted impacts against rigid surfaces, the $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ were significantly greater than those measured under all other conditions.

Conclusions. Vigorous shakes of this infant model produced rotational responses similar to those resulting from minor falls, but inflicted impacts produced responses that were significantly higher than even a 1.5-m fall onto concrete. Because larger accelerations are associated with an increasing likelihood of injury, the findings indicate that inflicted impacts against hard surfaces are more likely to be associated with inertial brain injuries than falls from a height less than 1.5 m or from shaking.

KEY WORDS • brain injury • child abuse • diffuse axonal injury • subdural hematoma • children

TRAUMATIC brain injury is the most common cause of death in children.³ Brain injuries resulting in hospitalization or death occur in at least 150,000 children per year, at a rate of more than 200 per 100,000 children. Head injury in infancy results in higher incidences of morbidity and mortality than those seen in older children, and it has become increasingly clear that the significant incidence of nonaccidental injury in the youngest patients is, in large part, responsible for this difference.^{5,11,28,29}

The majority of serious traumatic brain injuries in infants and toddlers is due to child abuse, and abused children with brain injury have a worse outcome than children who sustain an accidental brain injury.^{5,13} The actual mechanism of injury responsible for subdural hemorrhage, retinal hemorrhage, axonal injury, and skeletal trauma that characterize abusive head injury has been debated for decades. Caffey^{6,7} first proposed the term “whiplash shaken infant syndrome” to describe the occurrence of subdural and retinal hemor-

rhages in response to presumed inflicted angular acceleration of the head. Others have followed with reports of subdural hemorrhage, retinal hemorrhage, and death occurring in the absence of contact injury (skull fracture, cranial bruising, or scalp swelling).^{2,20} Regardless, contact head trauma remains a frequent finding in abusive head injury.¹⁷

Accidental falls are also a common cause of trauma found in the pediatric population, with falls accounting for 25 to 34% of hospital admissions for pediatric trauma and 6% of deaths in children due to trauma.^{21,34} Nevertheless, accidental falls are a common history given by caregivers in suspected abuse cases. The suspicion of abuse often arises when the event history appears not to correspond to the injury in the child. The differentiation between causes of accidental and abusive head injury is hindered by the controversy regarding fall heights associated with serious head injuries in children. Evidence exists to support the hypothesis that short falls do not cause serious injury and the critical height for a fall to cause death is substantial (> 10 ft).^{4,10,34,36,49,55} Simultaneously, however, others contend that relatively short falls can occasionally cause injuries associated with high mortality rates, such as SDHs, epidural hematomas, and skull fractures.^{8,18,19,21,40,43,44,46,54} Because the mechanical responses experienced by the head and the in-

Abbreviations used in this paper: ANOVA = analysis of variance; DAI = diffuse axonal injury; SDH = subdural hematoma; TAI = traumatic axonal injury; Δt = duration of the maximum change in angular velocity; $\Delta\dot{\theta}_{\max}$ = maximum (peak–peak) change in angular velocity; $\ddot{\theta}_{\max}$ = peak angular acceleration.

TABLE 1

Comparison between body measurements in infant and surrogate

Body Measurement	Infant	Surrogate
head weight (kg)	0.77–0.87*	1.13
head circumference (transverse plane [cm])	39.5†	40.5
head height (inferior to superior [cm])	13.5†	12.6
head breadth (right to left [cm])	10.5†	11.4
head length (anterior to posterior [cm])	14.1†	12.6
distance from shoulder to top of head (cm)	14.7†	15.1
distance from axis of rotation (C5–6) to top of head (cm)	15.4‡	14.7
distance from axis of rotation (C5–6) to center of gravity of the head (cm)	9.5‡	9.2
distance from axis of rotation (C5–6) to base of skull (cm)	3.3*	4.5
distance from axis of rotation (C5–6) to transducers (cm)	not applicable	18.0
total body weight (kg)	3–4* 4.8§	4.83
breadth of shoulders (cm)	17.6†	17.8
head/body weight ratio	0.23* 0.24	0.23

* Weight of 1-month-old infant according to Duhaime, et al., 1987.

† Measurements obtained in an infant 0 to 3 months of age according to Schneider, et al.

‡ Distance according to Swishchuk.

§ Weight of average male or female 6-week-old infant according to Kuczmarski, et al.

|| Ratio in a 6-week-old infant according to Jensen.

jury tolerances associated with shaking, shaking with impact, and falls have not yet been established, the differentiation between accidental and inflicted head injury is problematic. Objective information regarding these parameters is needed to determine if some circumstances are more hazardous than others during accidental and inflicted traumatic situations.

Previously, dolls that were anthropomorphically matched to human infants were shaken with and without impact;¹² investigators reported that during an inflicted impact the angular deceleration of the head is 45 times that observed during a vigorous shake, exceeding scaled thresholds for concussion, subdural hemorrhage, and DAI. In that study, however, the angular (rotational) motions were calculated on the assumption that there was a fixed center of rotation, and were not measured directly. Moreover, falls were not simulated, only inflicted shakes and impacts. In this study, we have created an improved anthropomorphic surrogate of a 1.5-month-old infant, and directly measured the rotational velocities experienced by the head of the surrogate during shakes, inflicted impacts, and short-distance falls. This research also extends that of previously published studies by examining the effect of different contact surfaces on the rotational response of the infant's head during both abusive and accidental injury scenarios. By comparing the responses, our findings are an important step toward distinguishing the potential for head injuries caused by rotational motion during contact and purely inertial events, and during accidental and inflicted injury scenarios.

Materials and Methods

Construction of the Anthropomorphic Dummy of an Infant

Head Design. A dummy was designed, constructed, modified, and

improved to create a more biofidelic and durable anthropomorphic surrogate of a 1.5-month-old infant than that used previously. The head of a toy doll (La Baby; JC Toys Group, Inc., Miami, FL) was used to represent the head of the anatomically correct 1.5-month-old anthropomorphic surrogate. Because the center of gravity and the majority of an infant's body mass are located in the head and torso, the distribution of the weight of the arms and legs of the infant were incorporated into the weight of the torso. The surrogate's total body weight, 4.8 kg (10.6 lb), was matched to that of a 1.5-month-old infant whose body weight lies within the 50th percentile.²⁵ Using previously reported measurements, the distributed masses of the head and body were adjusted to mimic those of 1.5-month-old infant by creating a head/total body weight ratio of 0.235^{12,25} (1.13-kg head mass). The breadth, length, and width of the head were measured and are in good agreement with those obtained in a 0- to 3-month-old infant in the 50th percentile (Table 1).

Neck Design. One-month-old infants have very compliant necks with little muscle tone and control of head movement.⁹ The weak flexor and extensor muscles of the neck allow a significant lag between the head and torso when raising the infant to a sitting position and lowering him or her back to a lying position.⁹ The normal movement of the neck has been described in the context of qualitative neurological examinations and developmental assessments in children; however, no quantitative information is available on the biomechanics of the human infant neck. In light of the absence of detailed quantitative information about the kinematics of infant necks in the literature, we fashioned a hinged neck with negligible resistance for the dummy, as previously published.¹² In this way, measurements would reflect a worst-case scenario of no resistance provided by the neck, so that we could ascertain the greatest possible velocities and accelerations that can be generated by these mechanisms. One end of a heavy-duty stainless-steel strap hinge was rigidly attached to the skull material on the surrogate's head; the other end was rigidly attached to the torso. The hinge was used as the neck joint in the surrogate to allow resistance-free motion in extension and flexion with no movement in other directions. The center of rotation of the hinge was located 9.2 cm inferior to the center of the mass of the head, and corresponded to the junction between C-5 and C-6 measured by magnetic resonance imaging.⁵¹ Although the fixed center of rotation in the dummy would result in an overestimation of rotational acceleration because the actual centers of rotation are likely to be higher in the cervical spine, the relatively short cervical spine, compared with the head size of a typical young infant, minimizes the influence of this idealization. Regardless, this simplification once again errs on the side of a worst-case scenario.

Skull and Scalp Material. Another design consideration was the representation of the stiff skull with a flexible scalp. Experiments were performed to determine appropriate "skull" and "scalp" materials to use in the construction of the dummy. The mechanical properties of the orthopedic-grade copolymer polypropylene (2.25 mm thick; American Plastics, Fort Worth, TX) were tested and determined to lie between those of infant skull and suture.^{31,42} The plastic was heated and molded to the head of the surrogate and allowed to cool to room temperature. The instrumentation mounting bracket was attached securely to the skull, creating a rigid connection among the instrumentation, surrogate skull material, and head. A latex rubber material (Mold Builder [1.25 mm thick]; ETI, Fields Landing, CA) with properties similar to scalp^{14,33,42} was used to cover the occipital portion of the polypropylene skull; this material remained adherent throughout the tests.

Head Instrumentation

An angular rate sensor (model ARS-01; ATA Sensors, Albuquerque, NM) was securely attached to the top of the dummy's head via a lightweight bracket, and was adjusted to measure rotations with an axis of rotation oriented perpendicular to the sagittal plane. This transducer location was selected to be remote from the impact site, so as not to damage the transducer in the impact and fall experiments. Any similar location on the head and transducer orientation would have yielded the same rotational velocity data. The velocity channel was sampled at 10,000 scans per second and filtered using a digital Butterworth low-pass filter (DADiSP/2000; DSP Develop-

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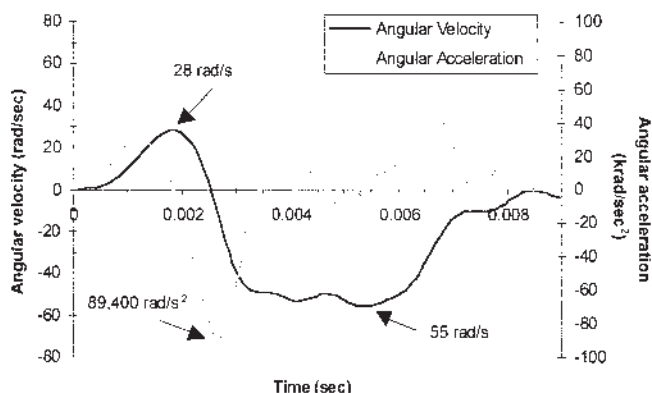


FIG. 1. Graph showing a representative angular velocity and acceleration trace for the impact produced from a 0.9-m (3-ft) fall onto concrete.

ment Corp., Newton, MA) as specified by the Society of Automotive Engineers.⁵⁰ Simulations of falls and inflicted impacts that involved contacting hard materials (carpet pad, concrete, or lab bench) were processed through a low-pass filter with a cutoff frequency of 1000 Hz, according to the standard SAE J211-1 channel frequency Class 1000 used for measuring head accelerations during automotive crash tests.⁵⁰ Because shakes and impacts against the foam mattress had pulse durations much longer than impacts against hard surfaces, these signals were processed through a low-pass filter with a cutoff frequency of 250 Hz to eliminate the noise from the velocity signal.

Abuse and Accident Reconstruction

A custom-designed drop-test apparatus was constructed to allow the dummy to be dropped consistently from 0.3-m (1-ft), 0.9-m (3-ft), and 1.5-m (5-ft) heights. The surrogate was suspended supine from three points (two on the body, one on the head) with the height of the dummy's head placed slightly lower (< 3 cm) than the body to ensure that the surrogate's occiput was the first point to contact the surface. To reproduce the shaking of an infant, the surrogate was grasped firmly by its torso and held at chest level. The dummy was then shaken back and forth, making sure that during each shake the head went through a complete range of motion from full extension to full flexion. Each episode included at least five shakes, with the final shake concluding with an inflicted impact of the surrogate's occiput against one of three materials located at approximately the volunteer's waist level (0.9 m from floor). The sequence was divided into two segments, the "shaking event" and the "inflicted impact event," for analysis. Volunteers were instructed to use maximum effort during vigorous shaking and impact, and not to release or throw the dummy during impact.

Three materials were used in the fall and inflicted impact simulations: a piece of 10.2-cm (4-in)-thick foam from a crib mattress, a section of 6.35-mm (0.25-in)-thick carpet pad, and a hard surface (concrete floor for falls, stone bench top for inflicted impacts). These materials were tested and the average linear elastic moduli of the foam and carpet pad were measured and found to be 24.8 and 621 kPa, respectively,⁴² whereas the elastic modulus of concrete in compression has been documented to range from approximately 20 to 60 GPa.⁴¹ These data demonstrate the wide range of contact surface characteristics that were used in the household fall situations.

Statistical Analysis

After filtering the measured angular velocity for each event, angular acceleration was calculated by taking the derivative of the angular velocity-time history trace. Only the rotation caused by the initial impact was analyzed in traces for falls and inflicted impacts. For each shaking event, only the shake with the greatest angular acceleration was analyzed out of the series of shakes in the episode. Each event was analyzed to find the $\Delta\dot{\theta}_{\max}$, the Δt , and the $\dot{\theta}_{\max}$ during this interval.

A total of 134 fall events were reconstructed from various heights

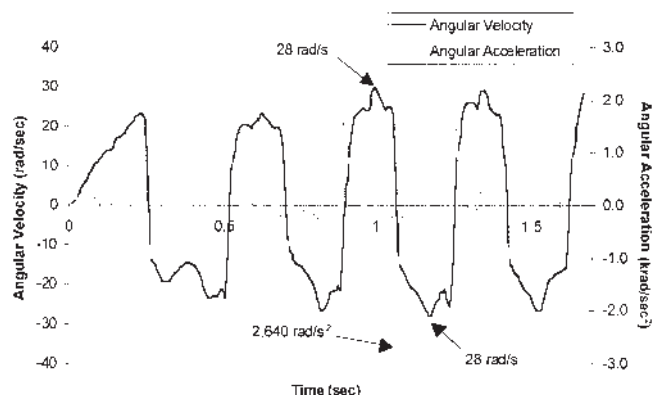


FIG. 2. Graph showing a representative angular velocity and acceleration trace for a shaking event.

onto different surfaces with at least 14 falls for each height-surface combination. A representative trace of the measured angular velocity and calculated angular acceleration during a fall is shown in Fig. 1.

Sixty-one shaking and impact sequences were performed by six different adult volunteers (four male and two female volunteers ranging in weight from 50–100 kg). Nearly all shaking episodes (60 of 61) consisted of a shake and an inflicted impact segment in which the surrogate's occiput made contact with one of three surfaces (foam in 18 episodes, carpet pad in 20 episodes, and a composite bench top in 22 episodes). Typical angular velocity and angular acceleration measurements from a series of shakes and inflicted impacts are shown in Figs. 2 and 3, respectively.

To determine overall differences between falls from different heights and onto different surfaces, three separate two-way ANOVAs were used to analyze the $\Delta\dot{\theta}_{\max}$, $\dot{\theta}_{\max}$, and Δt individually. A Tukey test for multiple comparisons was also performed to determine the differences between each individual type of fall. The measurements from shakes were compared with inflicted impacts against different surfaces by using one-way ANOVA to determine the significance of the test mode and the Tukey test to determine differences between each pair of test modes. A Dunnett test for multiple comparisons to a control was used to compare the $\Delta\dot{\theta}_{\max}$, $\dot{\theta}_{\max}$, and Δt from all falls, comparing the effects of the various surface materials by using shakes as the control or standard in the Dunnett test. The same analysis was repeated in a sequence of Dunnett tests by using inflicted impacts against foam, carpet pad, or bench top as the control group in each evaluation. Statistical significance was defined at a probability value of 0.05 or less for each analysis.

Results

Falls Onto Padded and Unpadded Surfaces

A two-way ANOVA revealed a significant overall increase in $\Delta\dot{\theta}_{\max}$ and $\dot{\theta}_{\max}$ ($p < 0.001$; Fig. 4 upper and center) and a decrease in Δt with falls onto harder surfaces and from greater heights ($p < 0.001$; Fig. 4 lower). A Tukey test revealed no significant effect of the height of the fall in the measured $\Delta\dot{\theta}_{\max}$ or $\dot{\theta}_{\max}$ during falls onto foam and showed that both kinematic measurements at a given height were significantly less for falls onto foam than for those onto carpet pad or concrete. Falls from a given height onto the carpet pad and concrete were indistinguishable, except for the $\dot{\theta}_{\max}$ during 1.5-m falls. For a particular surface, no significant difference between 0.9- and 1.5-m falls was found in the measurements of $\Delta\dot{\theta}_{\max}$ or $\dot{\theta}_{\max}$; however, measurements of Δt were significantly different when comparing fall heights onto foam, with the Δt decreasing as the height of the fall increased.

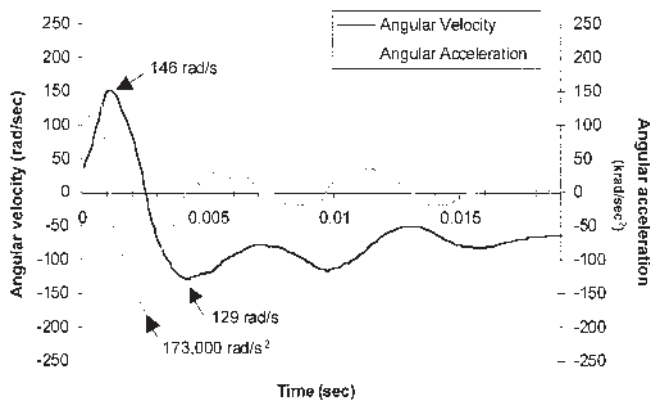


FIG. 3. Graph demonstrating representative angular velocity and acceleration trace for an inflicted impact against a bench top.

Shakes and Inflicted Impacts

Averaged across volunteers, the values of $\Delta\dot{\theta}_{\max}$ and $\dot{\theta}_{\max}$ during inflicted impacts against foam were greater than, although not significantly different from, those measured during shaking events (Fig. 5 *upper* and *center*). Inflicted impacts against foam, however, had a significantly shorter average Δt than shaking events (Fig. 5 *lower*). Inflicted impacts against the carpet pad and rigid bench surface were indistinguishable. Taken together, inflicted impacts against these two hard surfaces resulted in an approximately 39 times greater $\dot{\theta}_{\max}$, a three times greater $\Delta\dot{\theta}_{\max}$, and a 53 times shorter Δt than the response measured during shaking.

Shakes had a statistically similar $\Delta\dot{\theta}_{\max}$ to those of 0.3-m falls onto concrete and carpet pad, and a similar $\dot{\theta}_{\max}$ to that of falls onto a foam mattress. Shaking resulted in a significantly longer Δt than any fall events. Inflicted impacts against foam had a similar $\dot{\theta}_{\max}$ to that of falls onto a foam mattress and a similar $\Delta\dot{\theta}_{\max}$ to that of falls onto concrete. Inflicted impacts against a hard surface resulted in a similar Δt to those of falls onto carpet pad and concrete, and had a significantly greater $\Delta\dot{\theta}_{\max}$ and $\dot{\theta}_{\max}$ than all fall scenarios.

Discussion

The focus of this study was to determine the rotational response of the head of an infant that is experienced during low-height falls and inflicted head injuries. Rotational motions have been shown to cause a diffuse pattern of strains and injury throughout the brain, whereas translational motion causes more focal damage.³⁷ For this reason, with the exception of epidural hematoma, falls have often been considered benign, because they are assumed to be essentially translational events. Nevertheless, although a fall may have predominantly translational components, its terminus (contact with an often immobile object) may produce significant rotational events and the brain may also experience rapid changes in rotational velocity and deceleration. It was the purpose of this study to measure these rotational events and to compare them with those created by shaking, impacts after free falls, and inflicted impacts.

In the impacts against carpet pad and concrete simulated in this study, the occiput made contact before the torso and then rotated, producing a significant angular motion of the

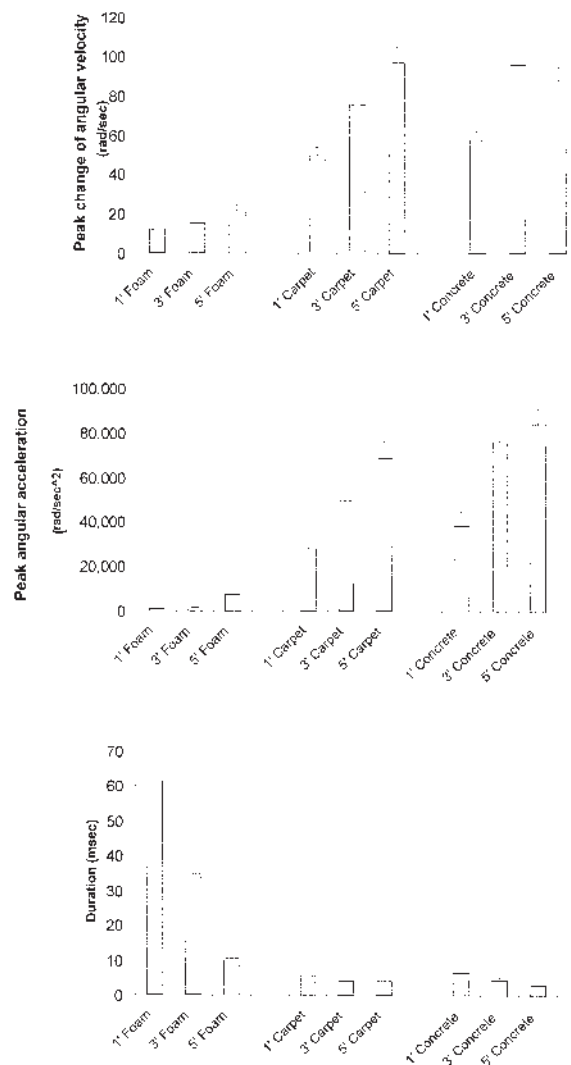


FIG. 4. Bar graphs showing the $\Delta\dot{\theta}_{\max}$ (*upper*), $\dot{\theta}_{\max}$ (*center*), and Δt (*lower*) for falls from different heights onto different surfaces. The bars depict mean values and the error bars indicate standard errors.

head relative to the thorax. We found that the rotational accelerations and changes in rotational velocity experienced by the head during impact increased with the height of the fall. These larger rotational responses can be attributed to the higher linear velocities reached as drop height (and thus potential energy) increases. Harder surfaces absorb less energy than deformable materials during contact, causing the head to rebound more; the head thus experienced significantly larger rotational accelerations during contact with the concrete and the carpet pad, compared with contact with foam, during falls and inflicted impacts. Conversely, when the head contacted the foam it was pocketed in the foam, such that both torso and head moved together, producing only a very small rotational response. It should be emphasized that the foam material used in these tests was uncased, and the addition of a plastic or other covering might alter the deceleration pattern. For this reason, one cannot extrapolate tests performed using uncased foam to impacts against a covered mattress or padded furniture.

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Head rotations during shakes occurred over significantly longer time periods than any other event. Thus, although shakes had a similar $\Delta\dot{\theta}_{\max}$ to that of a 0.3-m (1-ft) fall onto concrete and carpet pad, this change in velocity occurred over a much longer period, producing a significantly lower $\ddot{\theta}_{\max}$. Furthermore, shakes had a similar $\ddot{\theta}_{\max}$ to that of falls onto the mattress foam, but because the Δt is longer, shakes had significantly higher $\Delta\dot{\theta}_{\max}$ values.

There has been much debate on whether shaking alone is sufficient to cause the typical primary brain injuries seen in inflicted neurotrauma in infancy, specifically, SDH and/or TAI, or whether impact is necessary. Recent evidence suggests that injury to the cervicomedullary junction may be found in some cases of fatal inflicted head injury, and the role of this finding in the pathophysiology of apnea, hypoxia, and secondary cellular events is, at present, incompletely understood. Regardless, the focus of this study was to investigate the biomechanical causes of primary brain injuries by using rotational forces as a benchmark for the incidence of acute intracranial events such as failure of parasagittal bridging veins or axonal tears. To mimic inflicted impacts, the shaking scenarios described earlier were concluded with an impact of the surrogate's occiput against one of three surfaces. Our results demonstrate that measurements made during inflicted impacts against a carpet pad were not significantly different from those made during impacts against a rigid surface (lab bench top). Although the pad was less stiff than the bench top, the carpet pad used was only 6.35 mm (0.25 in) thick and the force of the inflicted impact completely compressed the pad during impact. During impact the carpet pad therefore exhibited the same properties as the underlying bench top during the later stages of contact. Inflicted impacts against carpet pad and bench top produced a significantly greater $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ than those experienced under all other conditions simulated in this study. In contrast, the inflicted impact against the thicker (10-cm) foam mattress did not fully compress the foam, resulting in a $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ that were lower than those associated with inflicted impacts against harder surfaces and similar to those associated with shaking. To summarize, inflicted impacts against the carpet pad and bench top had a three times greater $\Delta\dot{\theta}_{\max}$ and a 39 times greater $\ddot{\theta}_{\max}$ than shaking. These results suggest a higher likelihood of injury from inflicted impacts against hard surfaces than from vigorous shaking, or from falls of 1.5 m or less.

Sixteen years ago, we published results found using a less sophisticated anthropomorphic dummy of a 1-month-old infant,¹² and found similar $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ ratios between shakes and impacts to those in the current study. In the current study, we constructed a more biofidelic 1.5-month-old dummy by using skull and scalp material with properties similar to those in infants. We measured rotational velocity directly and expanded the study to include falls and a wide variety of contact materials. Despite these improvements, the current study has several limitations. The dummy designed for this study included a simplified representation of the infant skull and neck that potentially influenced loads during falls and impact events. First, the dummy's skull was made from a solid homogeneous 2.25-mm-thick sheet of copolymer polypropylene. The actual braincase of an infant consists of bone plates connected by compliant sutures that allow substantial deformation of the skull. The compliant skull deforms during the birthing process and permits nor-

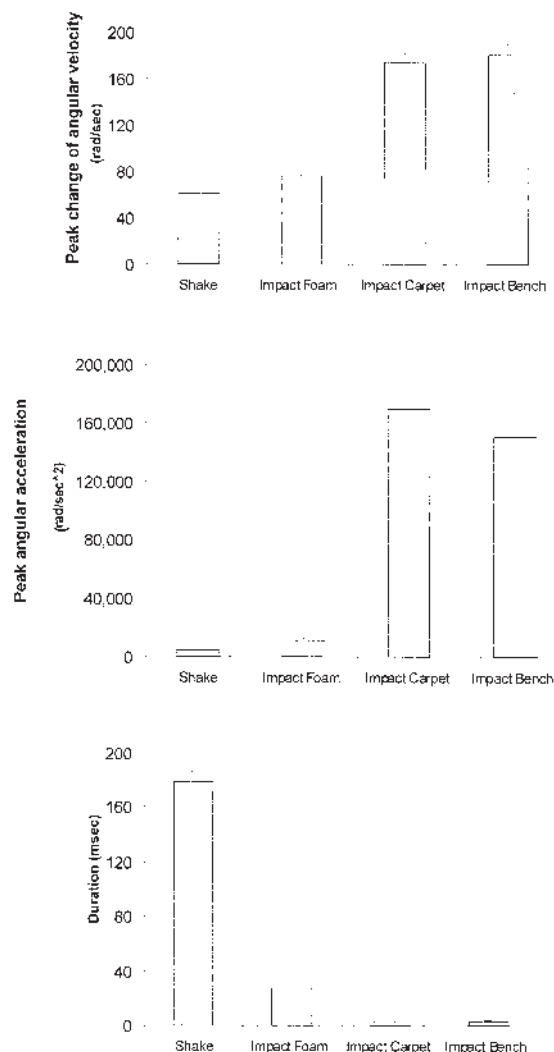


FIG. 5. Bar graphs showing the inflicted $\Delta\dot{\theta}_{\max}$ (upper), $\ddot{\theta}_{\max}$ (center), and Δt (lower) for shakes and impacts against different surfaces.

mal expansion of the brain and soft tissues during infancy and childhood; it also allows large changes in shape during impact loading.³¹ These skull deformations would slow the period of contact during an impact, and decrease the resulting angular accelerations. The solid plastic "skull" of the dummy is therefore not an exact representation of the separate bone and suture material of an infant, and it could cause an overestimate of the response measured during impact events. Studies including more accurate skull and suture representations, which can also be used to measure skull contact forces, are now underway; the important data obtained in these studies will be crucial for computational simulations of skull and brain injuries caused by these focal loads experienced during impacts.

A second limitation of the current study is the inability of the surrogate's neck to mimic the properties of a real infant's neck. At present, no detailed quantitative information is available to validate the biomechanical properties of the human infant neck. Qualitatively, as a child matures, the neck stiffness increases, offering more resistance to a rotational motion of the head relative to the torso. Nevertheless,

young infants (< 2 months old) have little muscle tone in their necks and cannot support the weight of their heads.⁹ Thus, the low-resistance hinge representation may be appropriate for a newborn but not an older child. Furthermore, the cervical spine consists of a series of vertebrae, allowing for rotation of the neck in different locations and directions; the hinge used to model the neck motion in the dummy has a fixed point of rotation, only allowing anteroposterior flexion and extension. The hinge does not provide resistance to motion or dampen responses between the torso and neck. More accurate kinematics of the infant neck would allow for translation as well as rotation of the head, and a moving center of rotation. These differences in the kinematics of the neck result in an overestimation of the rotational motions experienced during falls and inflicted events. Although the biofidelity of the surrogate has not been established, these experiments do provide an upper boundary for the measured head response to shaking, inflicted impact, and falls. When data are available on properties of the neck, a more biofidelic model should be created to ensure accurate measurements during inflicted and accidental injury scenarios.

A third limitation of the study is that the model represents a child in the 50th percentile for body and head mass. A heavier child with the same neck development would experience higher impact energy in a fall or inflicted impact due to the larger mass, and likely would experience a concomitant larger rotational velocity and acceleration than a smaller child. It is difficult to speculate, however, whether the volunteers could have generated the same peak acceleration with a heavier child. Considering that each sequence was a maximum effort, it is likely that the shakes would have resulted in a lower acceleration and velocity in a heavier dummy. Thus we would anticipate a greater disparity among the data obtained during shakes, impacts, and falls with increasing body and head mass.

Subdural hematoma and TAI are among the most common findings in serious head injuries in infancy and in those associated with nonaccidental causes. Both these injury types have been produced by, and correlated to, the angular velocity or angular acceleration of the head.^{1,15,16,32} Rotational motions have been shown to cause a diffuse pattern of strains and injury throughout the brain, whereas translational (linear) motion causes more focal damage.^{24,37} Although the anthropomorphic dummy test data are useful to evaluate the rotational response of the head caused by falls and inflicted injury events, the results of the dummy tests cannot be used to predict whether such rotations are sufficient to cause injury. Regional tissue thresholds specific to the infant would be required to predict injury on the basis of local intracranial stresses or strains produced by the rapid rotations. Such thresholds are currently unavailable for the pediatric population. In lieu of this information, we used a more qualitative approach to determine injuries likely to occur during simulated events. Specifically, we correlated measured accelerations and changes in velocity with injuries documented from controlled cadaver, animal, and human experiments in which the response is often measured directly and the exact details of the injury event are carefully recorded. Using dimensional analysis, angular velocities and accelerations from the different animal, human, and cadaver experiments were scaled to the infant as a function of brain mass (420 g).³⁸ These results were compared with the different rotational responses measured in the minor fall

and inflicted impact events simulated in the dummy experiments performed in this study.

Impacts from falls from 0.3, 0.9, and 1.5 m onto mattress foam, and from 0.3 m (1 ft) onto carpet pad produced the lowest rotational accelerations and changes in velocity of all falls examined in this study. Weber⁵⁴ found only a 10% chance of skull fracture when infant cadavers were dropped from a similar height onto a similar surface. No data have been collected from animal and human experiments that were conducted at the low levels of $\ddot{\theta}_{\max}$ and $\Delta\dot{\theta}_{\max}$ measured during falls onto foam. Values of $\ddot{\theta}_{\max}$ and $\Delta\dot{\theta}_{\max}$ similar to those measured during 0.3-m falls onto carpet pad were recorded from head rotations of instrumented models of boxers with no occurrence of concussion, skull fracture, SDH, or TAI of the brain.³⁹ All cases of SDH and DAI in human cadaver studies and primate (rhesus monkey and baboon) rotational inertial experiments^{1,15,32} had considerably greater angular accelerations and velocities when scaled to a human infant than an average $\ddot{\theta}_{\max}$ and $\Delta\dot{\theta}_{\max}$ produced by falls onto foam. Correlating these experimental data, it is highly unlikely that serious or fatal injuries occur during falls onto an unencased foam mattress from a height of 1.5 m or less, or from 0.3 m falls onto a carpet pad.

At the rotational responses calculated during impacts after 0.3-m (1-ft) falls onto concrete and 0.9-m (3-ft) falls onto carpet pad, it is not clear if serious injuries occur. The experimental evidence shows both the absence and occurrence of serious head trauma. Specifically, although human infant cadaver drop test studies have demonstrated an 80 to 100% occurrence of skull fractures,^{44,53} subhuman primate inertial studies^{1,15,32} and adult cadaver studies³⁵ have shown both the presence and absence of intracranial hemorrhage and acute SDH.

No experimental data could be found with a $\ddot{\theta}_{\max}$ and $\Delta\dot{\theta}_{\max}$ similar to those of the most severe falls, that is, 1.5-m (5-ft) and 0.9-m (3-ft) falls onto concrete and 1.5 m (5 ft) falls onto carpet pad. At lower values of $\ddot{\theta}_{\max}$, however, impacts to the cadaver head caused intracranial bleeding, and nonhuman primates experienced SDH and DAI. The absence of data at this highest tier underscores our uncertainty regarding the occurrence of serious injury at these levels. These falls represent an extreme limit, producing the maximum rotation because of the hinge neck, the occipital contact site, and the purely sagittal rotation. In reality a fall is likely to include mixed rotational directions that would decrease the actual acceleration in the sagittal plane and the potential for SDH. For these idealized situations of falls onto hard surfaces, experimental data thus support at least the possibility of intracranial injuries caused by these most severe falls measured in this study.

There were no experimental data at values of $\ddot{\theta}_{\max}$ and $\Delta\dot{\theta}_{\max}$ similar to those measured during shaking or inflicted impacts against foam. For example, shakes and inflicted impacts against foam each had a $\Delta\dot{\theta}_{\max}$ similar to that associated with noninjurious head rotation in boxers,³⁹ but these human tests produced a significantly greater $\ddot{\theta}_{\max}$. The most severe inflicted impact against foam approached, but was still less than, the values of $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ that produced SDH in adult rhesus monkeys and cadavers.^{1,15,32} To summarize, there are no data demonstrating that the $\Delta\dot{\theta}_{\max}$ and $\ddot{\theta}_{\max}$ experienced during shaking and inflicted impact against foam cause SDH or TAI in infants.

Finally, although the $\dot{\Theta}_{\max}$ and $\Delta\dot{\Theta}_{\max}$ measured during inflicted impacts against carpet pad and rigid bench top were significantly greater than those associated with all other scenarios tested in this study, no animal, human, or cadaver experiments at these levels of $\dot{\Theta}_{\max}$ have been published. The majority of inflicted impacts against these hard surfaces produced a $\Delta\dot{\Theta}_{\max}$ and $\dot{\Theta}_{\max}$ greater than the scaled rotational responses that produced fatal acute SDH in adult primates and intracranial bleeding in adult cadavers.^{1,15,35} Approximately half of these inflicted impacts also exceeded the scaled $\Delta\dot{\Theta}_{\max}$ and $\dot{\Theta}_{\max}$ that produced axonal injury in adult baboons.³² Given this experimental data, angular velocities and accelerations measured during inflicted impacts against hard surfaces would likely produce SDH and, possibly, TAI in an infant.

These injury projections should be interpreted with caution, because differences in species, age, material properties, geometry, and direction make scaling experimental angular acceleration and velocity measurements to infants problematic when based on differences in brain mass alone. To avoid the limitations of using scaled loads from animal and cadaver experiments to investigate real life events, case studies of minor falls in infants were also used to examine injuries that occur as a result of falling from different heights. Unfortunately, these falls are rarely witnessed, load measurements of the event are lacking, contact surface information is rarely given, and the population studied generally includes a broad age range, rather than just newborns. To increase the specificity of our comparisons, we included only case studies of children reported to be younger than 3 years old.

Skull fracture has been reported as a result of minor falls in children younger than 3 years.^{11,18,19,23,27,47,52,55} Reports of falls 0.6 m or less note an absence of skull fracture.^{44,48,53,55} Studies have documented cases of skull fracture from minor falls from heights estimated to be 0.9 m (3 ft)^{18,19,47,52} and 1.2 to 1.5 m (4–5 ft)⁵⁵ with no fatalities. Importantly, several studies have also shown an increase in the risk of skull fracture with the increased hardness of surfaces contacted after a free fall.^{27,54} To summarize, these case study data provide evidence that minor falls (< 1.5 m) can cause skull fracture, especially if the impact occurs on a hard surface.

Case studies of infants younger than 3 years old have shown that SDH, TAI, and death are rarely caused by impact from falls from 1.5 m (5 ft) or less. In a 3.5-year retrospective study, Chadwick and colleagues⁸ found no fatalities resulting from 1.5- to 2.75-m (5–9-ft) falls. In that study the authors did find three to seven deaths in children younger than 3 years of age with a history of falling from approximately 0.6 to 1.2 m (2–4 ft), but the majority of these cases had associated injuries attributed to abuse. Cases of playground falls reported by Plunkett⁴⁰ revealed only six fatal falls ranging from 0.6 to 1.8 m during an 11.5-year period. The causes of death in all six children were determined to be SDH or cerebral edema. Studies of falls occurring in hospitals from an estimated height of 0.61 to 1.2 m (2–4 ft) do not show any case of SDH or death.^{22,30,36} Other case studies have shown no incidence of SDH or death caused by falls from 0.9 m (3 ft).^{48,55} To our knowledge, no case study contains a report of serious brain injuries or death caused by falls from 0.3 m (1 ft) or less. Similar to the results of the animal and cadaver experimental data, these

data show that, although the possibility exists, SDH and death from minor falls are unlikely.

Conclusions

This paper presents the rotational response of the head of an infant, measured using an anthropomorphic dummy, during minor falls, shakes, and inflicted impacts. In general, the $\dot{\Theta}_{\max}$ and $\Delta\dot{\Theta}_{\max}$ increased with increasing fall height and surface hardness. The measured angular velocity and acceleration during minor falls were similar to those associated with shaking; however, inflicted impacts against hard surfaces produced a significantly greater $\dot{\Theta}_{\max}$ and $\Delta\dot{\Theta}_{\max}$ than a 1.5-m fall onto concrete. Because larger rotational acceleration and velocities are associated with a higher likelihood of injury, these findings suggest that inflicted impacts against hard surfaces may be more frequently associated with clinically significant inertial brain injuries than vigorous shaking or falls from less than 1.5 m. In addition, there are no data showing that the $\Delta\dot{\Theta}_{\max}$ and $\dot{\Theta}_{\max}$ of the head experienced during shaking and inflicted impact against unencased foam is sufficient to cause SDHs or primary TAIs in an infant.

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