Inertial Neck Injuries In Children Involved In Frontal Collisions

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ABSTRACT

There is a paucity of data regarding the potential for pediatric cervical spine injury as a result of acceleration of the head with no direct impact during automotive crashes. Sled tests were conducted using a 3-year-old anthropomorphic test device (ATD) to investigate the effect of restraint type and crash severity on the risk of pediatric inertial neck injury. At higher crash severities, the ATD restrained by only the vehicle three-point restraints sustained higher peak neck tension, peak neck extension and flexion moments, neck injury criterion (Nij) values, peak head accelerations, and HIC values compared to using a forward-facing child restraint system (CRS). The injury assessment reference values (IARVs) for peak tension and Nij were exceeded in all 48 and 64 kph delta-V tests using any restraint type. The test at a delta-V of 64 kph using only the vehicle belts as restraints resulted in peak upper neck tension, peak upper neck extension moment, and Nij values two times greater than the corresponding IARV. Only small differences were found in the injury metrics between a CRS installed with and without webbing tension except that head excursion was greater in the installation without webbing tension. These data show that the potential for neck injury exists for children involved in severe frontal crashes and restrained in either a forward-facing CRS or by vehicle belts–only, even in the absence of head contact.

INTRODUCTION

According to the National Highway Traffic Safety Administration (NHTSA), motor vehicle crashes are the leading cause of death in children ages 3 to 14 years old (NHTSA 2004). Child restraint systems (CRS) have been found to reduce fatal and serious injuries (Arbogast et al. 2004; NHTSA 2004; Elliott et al. 2006). However, in 2004, approximately half of the fatally injured children ages 1 to 4 years old were using a child safety system at the time of the crash. These data demonstrate that while current child restraint systems are effective in reducing fatal and serious injuries, a substantial number of fatalities occur to children restrained in child seats in motor vehicle crashes.

Currently the guidelines regarding child restraint usage from the American Academy of Pediatrics and NHTSA specify that a child ages from 1 to about 4 years old and weighing 20 to 40 pounds should be placed in a forward-facing CRS in the rear seat when riding in a motor vehicle. A child up to 80 pounds or more, between ages 4 and 8 years old, and shorter than 4 feet 9 inches tall should be seated in a belt-positioning booster in the rear seat of the vehicle. The majority of children involved in motor vehicle crashes are restrained by some means (Arbogast et al. 2004; Durbin et al. 2005). However, child restraint misuse and inappropriately restrained children have been reported at a significant rate (Bull et al. 1988; Hummel et al. 1997; Arbogast et al. 2004; Winston et al. 2004; Durbin et al. 2005; Brown et al. 2006). Improperly restrained children have been shown to have a significant increase in the severity of injuries (Brown et al. 2006).

While proper use of a child restraint has resulted in a significant reduction of the risk of head injuries, the risk of spinal injuries was not significantly reduced (Valent et al. 2002). Although cervical spine injuries are uncommon in the pediatric population, the mortality rate from these injuries is high (Givens et al. 1996; Patel et al. 2001). Motor vehicle crashes are a common cause of cervical spine injuries in young children (Givens et al. 1996; Zuckerbraun et al. 2004). Investigations using field accident data have found that the change of velocity (delta-V) of the vehicle during a crash is a strong predictor of injury risk for children (Brown 2006).

Currently the Federal Motor Vehicle Safety Standard for child restraint systems (FMVSS 213) does not explicitly incorporate any neck injury criteria. The standard currently requires child restraint systems (CRS) to be tested using peak sled accelerations of 19-25 g and a pulse duration of 75-90 ms. These requirements include only injury criteria for the head and chest. In 2002 NHTSA published a notice of proposed rule making (NPRM) that suggested amending FMVSS 213 to specify using the Hybrid III and CRABI ATDs with neck load cells and specifying a neck injury criterion (Van Arsdell 2005). The injury criteria proposed was the Nij criterion specified in FMVSS 208 without the peak tension and compression limits. The comments in response to this proposal mostly opposed the new neck
injury criterion. The respondents expressed concerns regarding increased cost and unintended consequences of instituting multiple, unevaluated changes to the standard, while the use of current CRS designs have resulted in extremely low injury rates despite extensive misuse. NHTSA conducted testing that showed many of the CRS currently on the market failed to meet this neck injury criterion, and no neck injury criterion was incorporated into the final rule. NHTSA suggested that the HIC36 requirement will “serve as a surrogate of sorts for a neck injury criterion.” (Van Arsdell 2005)

While neck injuries are uncommon in restrained pediatric occupants, case studies have been reported in the literature detailing restrained children who have sustained cervical spine and spinal cord trauma in frontal crashes without evidence of head contact (Fuchs et al. 1989; Huelke et al. 1992; Newman and Dalmotas 1993; Trosseille and Tarriere 1993; Weber et al. 1993; Stalnaker 1993). Many of these cases involved young children (< 6-years-old) involved in severe or moderate severity frontal crashes (delta-V ≥ 45kph) and restrained in various configurations including lap belt only, lap-and-shoulder belts, and a variety of CRS systems. The injuries consisted of fractures, dislocations, and/or spinal cord damage in the high cervical spine. The postulated mechanism for these types of injuries is inertially-induced loading of the cervical spine during these crashes (Huelke et al. 1992; Newman and Dalmotas 1993). Additionally, the authors have personally investigated a number of other frontal collisions in which children restrained in a variety of configurations (including vehicle restraints only, forward-facing CRS, and booster seats) have sustained cervical spine injuries without any evidence of head contact.

The anatomy and anthropometry of the young pediatric population make them particularly vulnerable to inertial neck injuries. As observed in a number of previous studies, children are not miniature adults. Young children have a relatively large head and smaller neck (Burdi et al. 1969; Huelke 1998). Also, the pediatric spinal ligaments are more lax and the neck musculature is less developed, allowing for increased motion of the cervical spine (Fuchs et al. 1989). The vertebral bodies are not fully ossified and the facet joints are more horizontal than in adult spines (Fuchs et al. 1989). All of these anatomical characteristics of children make them more susceptible to inertial neck injuries in frontal collisions.

The purpose of this study is to investigate the biomechanical response of children during frontal crashes without head contact. One aim of these tests is to systematically examine the effect of crash severity on the potential for pediatric inertial neck injuries. Also, this study examines the injury risk to a child properly restrained (forward-facing CRS installed properly), misusing a restraint (forward-facing CRS installed without recommended vehicle webbing tension), and using an age-inappropriate restraint configuration (three-point vehicle belts only).

METHODS

To investigate the potential for inertial neck injuries, sled tests were conducted using anthropomorphic test devices (ATD) in different restraint configurations at different crash severities. Tests were conducted using a pneumatic acceleration system, a programmable wire-bending decelerator, and a 170-ft rail that minimizes test article disturbance during the acceleration phase (Seattle Safety LLC). Tests were conducted at 32, 48, and 64 kilometer per hour (kph) change of velocity for each restraint configuration.

A 1996-2004 model Chrysler minivan second row bench seat was rigidly secured to the sled. The bench seat and buckle stalks were replaced with the same type of seat after each severe (64 kph) test. The vehicle seatbelt restraint system and corresponding hardpoints used in each tests were from the Chrysler minivan. The seatbelts consisted of a three-point continuous loop system with a vehicle-sensitive and webbing sensitive, emergency-locking retractor. The system did not include an automatic-locking feature. The entire restraint system (belt, retractor, anchors), with the exception of the buckle stalk, was replaced by an unused system prior to each test.

ANTHROPOMORPHIC TEST DEVICES AND RESTRAINT CONFIGURATIONS

A 3-year-old Hybrid III ATD was placed on the right outboard position of the bench seat. The ATD was secured in three different restraint configurations at the three different impact severities. For the first configuration, the ATD was placed in a properly installed forward-facing CRS. A new model CRS [Titan, Model# 3671595P1 Evenflo Company, Inc.] was used for each test in this series. The model of CRS used during these tests weighed 10.4 pounds and was chosen based on its high rating by Consumer Reports (May 2003). In the first configuration, the CRS was installed according to the manufacturer’s instructions. The installer applied his body weight to the seat and then tightened the vehicle restraints so that there was tension in the webbing. Because the vehicle restraint system did not include an automatic-locking feature, the vehicle belts were routed through the CRS and secured using the manufacturer supplied locking clip according to the instruction manual. Tethers were not used in any of the sled tests conducted as part of this study. To investigate a common misuse of a CRS, the second configuration used the same model forward-facing CRS installed according to the manufacturer’s instructions, except without applying tension to the webbing. The vehicle belts were positioned such that the webbing had a minimal amount of tension while not allowing any slack to occur between the belt and CRS. Again, the CRS was replaced by a new unused CRS at the beginning of each test. For each CRS configuration, the location of the latchplate and locking clip was measured relative to the vehicle restraint anchor position. Similarly, the location of the CRS belts and chest clip were measured relative to the
CRS. These locations were the same for each test to ensure the repeatability of the CRS and ATD installation between tests. The third configuration examined an inappropriate restraint configuration for a 3-year-old with the ATD placed on the bench seat without the CRS and restrained only by the vehicle restraints.

In each of the same tests, a six-year-old Hybrid III ATD was placed on the left outboard position in 3 different restraint configurations. The ATD was placed in a booster seat [TurboBooster, Model# 8498SRT Graco Children's Products Inc.] and restrained with the vehicle belt system for the first configuration. The booster seat was replaced with a new seat before each test. This model booster seat was chosen based on its high rating by Consumer Reports (May 2003). For the second configuration, the ATD was placed on the bench seat and restrained by the vehicle belts with the shoulder belt placed behind the torso of the ATD. The third configuration consisted of the ATD seated on the bench seat restrained by the vehicle belts with the shoulder belt placed across the torso. The vehicle restraints were replaced with an unused restraint prior to each test. The results from the 6-year-old ATD are beyond the scope of this communication and will be presented in a subsequent paper.

INSTRUMENTATION

The impact velocity of the sled was measured using a velocity trap and the sled acceleration was measured using sled-mounted accelerometers. Shoulder and lap belt loads were measured during each test. High-speed video was recorded using an on-board digital camera on the right and an off-board digital camera on the left side. Target markers were placed on the rigid sled frame, on the CRS (when used), and at the approximate center of gravity of the ATD head as viewed from the side.

Head acceleration was measured with three uniaxial accelerometers mounted at the head center of gravity. Six axis load cells measured neck loads, one at the upper neck and the other at the lower neck location. Chest acceleration was measured using three uniaxial accelerometers and chest deflection was measured using a potentiometer.

DATA ANALYSIS

Each channel of data was recorded at 10,000 samples per second and filtered according to crash test standards [SAE J211]. The high-speed digital video was recorded at 500 frames per second.

The head acceleration data and neck load data were only analyzed for the inertial pulse prior to contact with any structure, including contact between the head and other parts of the ATD. Peak resultant acceleration, $HIC_{15}$, and $HIC_{36}$ were calculated from the head acceleration data acquired. Head excursion from the initial seated position was measured from the high-speed video using Image Express (Sensors Applications Inc., Utica, NY). Tensile load, peak neck moment and neck injury criterion ($Nij$) were calculated using the data obtained from the upper neck load cell. The $Nij$ was calculated according to the FMVSS 208 specification for the 3-year-old ATD.

RESULTS

The results for each of the nine tests conducted are reported in the appendix in Table A1 and discussed below.

CRASH SEVERITY

The measured delta-V of each sled test was within 1 kph of the target crash severity. The average peak sled accelerations for the 32, 48, and 64 kph tests were -16.6, -25.2, and -33.6 g respectively. The pulse durations of each test were similar with the average pulse time of 96 ms. The shape of the acceleration-time history was approximately a haversine shape (Figure 1).

![Figure 1: Typical sled accelerations. Tightly installed CRS tests at 3 different severities.](image)

UPPER NECK LOADS

The peak axial loads measured in the upper neck were tensile forces for all tests (Figure 2). Increased crash severity generally resulted in increased peak upper neck tension for each restraint configuration. The 48 kph tests produced peak tension values greater than the 32 kph tests for each type of restraint. The peak tension values for the 48 kph and 64 kph tests were comparable for a tightly installed CRS. In contrast, the tension measured during the 64 kph belts-only test was 35% greater than the 48 kph test using the same restraint configuration.
Upper neck peak tension varied between the different restraint configurations tested (Figure 2). The belts-only 32 kph test resulted in decreased upper neck tensile load compared to the tight and loose CRS installations. In contrast, the upper neck tension in the 64 kph belts-only test was greater than the CRS configurations. The CRS installation without webbing tension produced upper neck tensile forces slightly lower than the tight installation at 32 and 48 kph. Tensile loads were comparable for both CRS tests at 64 kph.

The peak upper neck moments measured in the 3-year-old ATD were all extension moments. The peak extension moments increased with increasing crash severity for the belts-only restraint configuration (Figure 3) and were comparable for all crash severities in both CRS configurations. The peak neck moments were similar across delta-V for both CRS tests. For the 48 and 64 kph tests, the peak moment was more than double the value recorded during the CRS tests.

The peak flexion moments in the upper neck were greater in the belts-only tests when compared to the CRS tests at the same crash severity (Figure 4). The flexion moment during the belts-only, 32 kph tests was greater than that in any accident severity in both CRS restraint configurations.

UPPER NECK INJURY ASSESSMENT REFERENCE VALUES

Peak tensile load

One of the neck injury assessment reference values specified in FMVSS 208 for the 3-year-old dummy is a peak tensile upper neck load of 1130N. Eight out of the nine tests conducted in this study resulted in peak upper neck loads greater than this reference value (Figure 2). Only the belts-only 32 kph test resulted in peak upper neck tension less than this criterion (1084N).

Neck Injury criterion - Nij

The injury criteria Nij was calculated for each sled test using the methods and critical values specified for the 3-year-old ATD in the federal motor vehicle safety standards. The maximum value of Nij for each test was in tension and extension (Nte). The maximum Nij values during the 32 kph tests for the CRS and belts-only configurations were approximately 1 (Figure 5). The 32 kph, belts-only test was the only test to be below the Nij criterion.
The peak Niij value increased slightly with increased crash severity for both CRS configurations. For both configurations, the Niij increased from approximately 1.1 in the 32 kph CRS tests to 1.35 in the 64 kph tests (Figure 5). Also, the relative contribution of the upper neck tension increased slightly with increased delta-V in the CRS tests. The tension component of peak Niij was approximately 60% for both 32 kph CRS tests and increased to 75% when the delta-V was increased to 64 kph.

There was a more dramatic increase in the peak Niij with increased delta-V for the belts-only configuration (Figure 5). The peak Niij more than doubled between the 32 and 64 kph belts-only tests. The tension component of the peak Niij was approximately 50% for the 32 kph and 48 kph belts-only tests. The contribution of tension decreased to 28% of the peak Niij during the 64 kph belts-only test.

The peak Niij values for all restraint configurations were similar at the 32 and 48 kph crash severities. The peak Niij values ranged between 0.8 and 1.0 for each of the restraints used at 32 kph. The 48 kph tests produced a peak Niij of between 1.1 and 1.45 for each restraint configuration. Both CRS tests at 64 kph also showed similar peak Niij values (1.35). The peak Niij value for the belts-only 64 kph test was greater than the CRS tests at the same crash severity. This test resulted in a peak Niij of 2.0, exceeding both the tension and extension component of the Niij criterion at the same point in time. The belts-only tests showed a lower contribution of tension to the peak Niij value at each crash severity compared to the CRS tests. The most dramatic difference was during the 64 kph tests where the neck tension comprised approximately 75% of the peak Niij for the CRS tests and only 28% for the belts-only configuration.

The CRS configurations at each crash severity had minimal flexion moments of the upper neck (Figure 4). The tension-flexion neck injury criterion from each of these tests was below 0.35.

**LOWER NECK LOADS**

The peak tension in the lower neck load cell increased with increased crash severity for each restraint configuration (Figure 6). All configurations at 48 kph were greater than all configurations at 32 kph. Similarly, all tests at 64 kph were greater than all configurations at 48 kph. Within each restraint configuration the peak tension more than doubled between the 32 kph and 48 kph tests. The 64 kph peak tension increased 55%, 29%, and 47% over the 48 kph sled test within the CRS tests with and without webbing tension and belts-only configurations respectively.

At each crash severity, the peak lower neck tension varied between the different restraint configurations. At each severity, the peak tension for the belts-only configuration of the 3-year-old was greater than both the CRS configurations. The lower neck tension in the belts-only configuration increased between 30% and 50% over the CRS configurations for the three crash severities tested. The two CRS configurations resulted in similar lower neck loads at each crash severity.

**LOWER NECK MOMENTS**

During all sled tests conducted, the peak moments in the lower neck were flexion moments and the extension moments were minimal. For the belts-only tests, the peak lower neck moment increased with increased crash severity (Figure 7). In contrast, the peak moments were similar in both CRS configurations across all crash severities, with slightly lower values in the CRS installed without webbing tension.
The peak moments measured during the 32 kph CRS tests were greater than the belts-only test. However, the 64 kph tests resulted in greater peak moments in the belts-only configuration when compared to the two CRS installations. The moments in the 48 kph tests were similar across restraint configurations.

HEAD ACCELERATIONS

Noise spikes were observed in the longitudinal head acceleration (Ax) data in the CRS tests with webbing tension and the 32 kph and 48 kph belts-only test. This accelerometer then failed after the crash pulse during the 48 kph belts-only test and had to be replaced. These noise spikes were confirmed after reviewing the video, which demonstrated that no impact to the head corresponded to these short duration spikes. These spikes were disregarded during the peak acceleration and HIC analyses. HIC values varied less than 3% between the data with and without the noise spike.

For the belts-only tests, the high-speed video revealed that the ATD head contacted the upper and/or lower extremity during its forward movements. This was confirmed with the head acceleration data that showed a short duration pulse corresponding to the head contact. This contact occurred after the peak inertial head acceleration pulse. These head impact pulses were not observed in the head acceleration data traces for the CRS tests. The purpose of this study was to examine the response of the ATD during the inertial pulse without any head contact. Therefore, the HIC values for the belts-only tests were computed only up until the occurrence of head impact with the extremities.

Peak resultant linear acceleration of the head increased with increased crash severity for each restraint configuration (Figure 8). Within the same restraint configuration, the peak accelerations approximately doubled between the 32 kph to the 64 kph tests. The peak head acceleration measured during the belts-only configuration was higher than the CRS tests for each crash severity. The test using belts-only resulted in 30% to 50% greater head accelerations than both CRS configurations.

HEAD EXCURSIONS

The peak horizontal excursion of the ATD head differed between the restraint configurations (Figure 9). The CRS tests without webbing tension resulted in increased head excursion over the CRS tests with webbing tension and belts-only tests at the same crash severity. The belts-only tests had smaller excursion than the CRS tests at each crash severity. For all restraint configurations, the excursion of the head increased with increased delta-V.
HEAD INJURY ASSESSMENT REFERENCE VALUE

The federal safety standards (FMVSS 208) and other researches have defined the head injury criterion as a HIC$_{15}$ value of 570 for the 3-year-old ATD (FMVSS 208, Mertz et al, 2003). The HIC$_{15}$ measurements were below this reference value for all 32 kph tests (Figure 10). Also, the 48 kph tests using the tight and loose CRS restraints resulted in HIC$_{15}$ values below this value. However the 48 kph, belts-only test resulted in a HIC$_{15}$ value greater than the reference value of 570. All restraints at 64 kph exceeded the reference value. The 64 kph, belts-only test resulted in a HIC$_{15}$ value more than three times greater than the injury reference value.

The head injury criteria HIC$_{15}$ and HIC$_{36}$ increased with increased crash severity (Figures 10, 11). Within the same restraint configuration, the HIC$_{15}$ values for the 48 kph tests were more than three times greater than the 32 kph tests. The CRS configurations produced an increase of approximately 50% between the 48 and 64 kph tests. The belts-only tests showed an increase of almost 150% between these two severities. Similar increases in HIC$_{36}$ were noted with increasing crash severity.

The belts-only configuration resulted in a greater HIC$_{15}$ value than the CRS configurations for each crash severity (Figure 10). As the delta-V increased this difference increased. The HIC$_{15}$ values during the 32 kph tests are similar for all restraints. At 48 kph, the belts-only test resulted in a value more than 50% greater than the HIC$_{15}$ determined for CRS configurations at the same delta-V. The belts-only test at 48 kph produced a HIC$_{15}$ value comparable to the CRS tests performed at an increased delta-V of 64 kph. The HIC$_{15}$ value for the belts-only test at 64 kph was more than double the value calculated for the CRS tests at the same severity. A similar increase in the 64 kph belts only configuration was also observed in the HIC$_{36}$ data (Figure 11).

CHEST ACCELERATION

The 3 ms clip peak chest resultant acceleration increased with increasing delta-V for each restraint configuration (Figure 12). The peak acceleration increased approximately 60% between the 32 and 48 kph tests for each restraint configuration. The CRS tests produced only an approximate 15% increase from 48 to 64 kph. However, the belts-only configuration produced an increase of 45% between the same severities.
During the 32 and 48 kph tests, the belts-only test resulted in slightly lower chest accelerations than both CRS configurations. The peak acceleration during the 64 kph tests were similar between the restraint configurations. The peak accelerations measured during the 48 kph were greater than all tests conducted at 32 kph. Similarly, the 64 kph tests for all restraint configurations resulted in accelerations greater than all the 48 kph tests.

CHEST DEFLECTION

As the crash severity increased, the peak chest deflections also increased within each restraint configuration (Figure 13). The peak chest deflections were similar for the two CRS restraint configurations at each crash severity. The belts-only configuration resulted in more chest deflection than the CRS restraints at each delta-V.

CHEST INJURY ASSESSMENT REFERENCE VALUE

The federal safety standards define the injury reference value for chest injury to be an acceleration (3 ms clip) of 55 g for the 3-year-old ATD. This value was exceeded during the 64 kph tests for all restraint configurations. The peak chest acceleration during the 48 kph test using the CRS installed without webbing tension was also greater than the reference value.

Mertz et al (2003) defined the reference value for peak chest deflection as 28mm for the 3-year-old ATD loaded by a shoulder belt. This value was only exceeded during the 64 kph test with the belts-only restraint configuration.

DISCUSSION

The purpose of this study was to examine the inertial neck loads produced in a 3-year-old ATD during frontal crashes at different severities and using various restraint configurations. Peak upper and lower neck tension, peak upper neck extension moments, peak upper neck flexion moments, and Nij values were higher for the ATD restrained by only the vehicle belts compared to using a forward-facing CRS. This demonstrates that of the configurations tested, the 3-year-old restrained in the vehicle seatbelts without a CRS has the highest risk of neck injuries caused by inertial (non-contact) conditions. Lower neck loads were less than upper neck loads, consistent with anecdotal field accident data showing more upper neck injuries (Fuchs et al. 1989).

Only small differences were found in the ATD response between a CRS installed with and without webbing tension, except for head excursion. During the tests of the improper CRS installation, the CRS was not coupled to the vehicle seat during the initial phase of the crash pulse due to the lack of tension in the vehicle belts. Therefore during this initial phase of the deceleration, the CRS moved relative to the seat and basically increased the effective mass of the ATD (ATD 35 lbs, CRS 11 lbs). However, this effect was minimal resulting in only small differences in the injury metrics. For the 3-year-old, this study demonstrates that a properly installed forward-facing CRS restraint configuration minimized the potential for head and neck injury. However, in every test at 48 and 64 kph delta-V, irrespective of restraint type, the injury reference values for peak upper neck tension and Nij were exceeded. This demonstrates that the potential for inertial neck injuries exists during severe frontal crashes even for properly restrained children.

The CRS installed without webbing tension resulted in larger head excursions than the CRS with webbing tension. The difference between head excursions in the two configurations was approximately five to ten centimeters. The addition of a top tether would likely reduce head excursions in the CRS configurations. Additionally, research using ATD and cadaver data has shown increased head excursions in the cadaver test compared to a test conducted using the same crash
parameters and a 3-year-old ATD (Wisman et al. 1979). Any increase in total head excursion will increase the risk of head contact with forward structures. In the absence of contact, injuries are not caused by excursion. Injuries are the result of relative motion between the child’s head and torso as the torso is decelerated by its interaction with the restraint system. This is shown in the fact that the CRS installed without webbing tension resulted in increased head excursions but comparable head and neck loads to the CRS installed with webbing tension. Because of their anthropometry with the head mass being a larger portion of the total body mass and their relatively less developed neck anatomy, children will by their nature be more susceptible to inertial neck injury. Any attempt to use technologies such as force-limiters to decrease the ride-down accelerations of the torso will necessarily increase the excursion of the child occupant, increasing the risk of impacting interior vehicle structures.

Increased head accelerations in the belts only configuration were associated with increased neck loads in that configuration, compared to both CRS configurations (Figure 8). Chest accelerations were comparable for all three configurations (Figure 12), suggesting that all configurations provided similar restraint at the mid-torso, where the accelerometer is located. However, differences in upper torso restraint due to geometrical differences between the CRS (both shoulders restrained) and vehicle belt restraint (one shoulder restrained) configurations affect the upper torso kinematics, as well as the motion of the head and neck. The decreased neck loads measured in CRS tests could also be a result of the added compliance of the CRS structure and CRS belts. This increased compliance allows for decreased ride-down accelerations and increased excursion of the ATD when restrained in a CRS (Figure 9).

Other researchers have also performed sled testing using the 3-year-old ATD. To our knowledge, no other testing has been performed at speeds as high as the 64 kph delta-V done in this series, nor has there been a study evaluating the effect of vehicle restraint webbing tension using instrumented ATDs. Henderson et al. (1997) performed 48 kph sled tests with a 3-year-old ATD restrained in three-point restraints. They reported neck tensile forces and HIC values comparable to those reported in this study; however, they measured upper neck flexion moments several times larger than those reported in the current study. Kapoor et al. (2006) performed a sled test at 48 kph using a 3-year-old ATD restrained in a forward-facing CRS and reported lower values of upper and lower neck tension, Nij, and HIC15, compared to the present study. Values for neck moments and HIC15 were comparable. Menon et al. (2004) performed a series of sled tests using a 3-year-old ATD in three restraint configurations, including a forward-facing CRS and vehicle belts only. While these tests were performed at different velocities than the current study, preventing direct comparisons with the current work, trends in the two studies could be compared. For most measurements, the two studies reported similar trends. For example, in both studies, Nij increased with increased crash severity. Nij values were higher for the belts only configuration than the forward-facing CRS configuration in both studies. However, Menon et al. reported higher head accelerations in the forward-facing CRS configuration compared to vehicle belts only; we observed the reverse.

NHTSA published an NPRM that suggested amending FMVSS 213 to incorporate a neck injury criterion. The injury criterion proposed was the Nij criterion specified in FMVSS 208 without the peak tension and compression limits. The comments in response to this proposal mostly opposed the new neck injury criterion, and no neck injury criterion was incorporated into the final rule. A HIC36 criterion of 1000 was used to help reduce the potential for neck injury. Interestingly, the results of the current study show that HIC36 only exceeded 1000 in the 64 kph tests, while Nij exceeded 1.0 in several tests where the HIC36 value was below the threshold. Therefore, it is possible that there are mechanisms of neck injury that are not captured by the HIC36 criterion.

Due to the complexity and variation of vehicles, vehicle restraint systems, child restraint systems, and their occupants, it is impossible to examine all aspects of child restraint systems in a single series of tests. The sled tests conducted were limited to frontal crashes only, a configuration chosen because frontal collisions have been identified as a major source of cervical spine injuries to child occupants. Also, the crash pulse used for these tests had the same duration for each severity tested. This pulse does not represent all real-world crashes and the effects of pulse duration and pulse shape were not examined. Test-to-test variability was minimized by using a repeatable sled crash pulse, installing new restraints (vehicle and CRS) for each test, and replicating the vehicle and CRS belt geometry for each restraint configuration. However, each restraint configuration was tested only once at each crash severity therefore no quantitative information is available on the variability of each test. Because this study focused on inertial (non-contact) injuries only, a front seat was not present during these frontal tests. However, the presence of a front seat could lead to head contacts during real-world crashes and associated changes in head accelerations and neck loads.

Only one type of child restraint was used during the sled tests. The consequences of different models or brands of CRS were not examined. Lower anchors were not present in the bench seat used for the sled tests. Therefore, the LATCH (Lower Anchors and Tethers for Children) system was not evaluated in this series of tests. The effects of not installing a CRS with the recommended tension in the vehicle belts were examined in this study. However, various types of restraint misuse have been documented, which have different effects on occupant kinematics and loading. The consequences of each misuse scenario were not investigated.
The issue of biofidelity of the child dummies has been extensively discussed in the literature (e.g. Wisman et al. 1979; Brun Cassan et al. 1993; Menon et al. 2004). Because of the paucity of data on the mechanical properties of the pediatric cervical spine, relative to that available for adults, identifying tolerance values for children is difficult. Much of the pediatric data used in the current FMVSS standards were obtained by scaling adult tolerance values and from animal models (Backaitis et al. 1975; Irwin and Mertz 1997; van Ratingen et al. 1997; Pintar et al. 2000; Ching et al. 2001; Hilker et al. 2002; Nuckley et al. 2002; Mertz et al. 2003). Research to obtain data from pediatric cadaver testing and other sources is ongoing (e.g., Nuckley et al. 2005; Ouyang et al. 2005; Prange et al. 2004) and should be used as it becomes available.

Our sled test series demonstrated a substantial increase of the peak Nij value with increased crash severity regardless of restraint configuration. These results suggest that all children involved in severe frontal collisions have a risk of sustaining a serious cervical spine injury. Analyses of real-world data (Valent et al. 2002; Zuckerbraun et al. 2004) have not shown significant rates of neck injuries for children involved in automotive crashes. However, many of these analyses do not examine the severity of the crashes. As shown in our study, neck injury potential increases substantially with increased delta-v. Because the occurrence of crashes with frontal delta-V’s of 48 to 64 kph involving children are rare in the real-world, the sample size of these field accident analyses might not be large enough to capture many of these injuries.

In our test series, the 64 kph delta-v tests resulted in a peak Nij of 1.35 for the CRS restraints and 2.0 for the belts-only configuration. The risk of serious (AIS 3+) neck injury for these Nij values is approximately 25% and 95%, respectively (Mertz et al. 2003). Henderson et al. found no instances of cervical spine injuries for properly restrained children in a study of 62 children involved in frontal collisions (Henderson et al. 1994). They reported the absence of neck injuries in 4 cases of children properly restrained in forward-facing child restraints at delta-Vs of over 60 kph. This could be the result of the limited number of cases investigated and the 25% risk of serious neck injury associated with the properly restrained ATD in our severe sled test. Henderson et al. did report a fatal neck injury of a child improperly restrained by a vehicle’s three-point belt during a 65-70 kph delta-V crash. This is consistent with the high risk of neck injury associated with an Nij value measured during our sled test at 64 kph, belts-only configuration. Our data indicate that if a child is involved in a rare, severe frontal collision, there is a risk of inertially-induced cervical spine injury.

CONCLUSIONS

A series of frontal sled tests was performed using a 3-year-old ATD seated in three different restraint configurations: a properly installed CRS, an improperly installed CRS, and using the vehicle belts only (no CRS). ATD injury measurements increased with increased crash severity. The belts only configuration produced the highest neck tensions, neck moments, Nij values, head accelerations, and HIC values. With the exception of head excursions, the amount of vehicle belt webbing tension used to install a CRS did not substantially affect head accelerations and neck loads.

REFERENCES


Henderson, M., J. Brown et al. (1994) "Injuries to restrained children." Association for the Advancement of Automotive Medicine, 38th Annual Conference, Lyon, France, AAAM.


CONTACT

Dr. Michael T. Prange has a Ph.D. in Bioengineering from the University of Pennsylvania and currently works for Exponent. He can be contacted at (215) 594-8800 or mprange@exponent.com.
### APPENDIX

<table>
<thead>
<tr>
<th></th>
<th>CRS with tension</th>
<th>CRS without tension</th>
<th>Vehicle belts only</th>
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<tr>
<td></td>
<td>32 kph</td>
<td>48 kph</td>
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Table A1: Results from the sled tests using 3 different restraint configurations and 3 different crash severities.
More evidence that shaking alone does not result in head injuries and that neck injuries are more likely to occur than head injuries in shaking without impact.

Just to put the injury tolerance levels and the loads during shaking into perspective:

I recently published a paper (see attached) where we ran frontal crash simulations using a pneumatic sled and measured the head and neck response of a 3-year-old dummy restrained at the torso by the 5-point belt of a forward-facing child car seat. So, this is a situation analogous to shaking where the torso is held and the head rotates about the neck without contact. Because we simulated a frontal crash and the child seat was faced forward, the deceleration created flexion of the head (chin-to-chest).

During a 20mph crash, the resulting head accelerations were 40-45g, well below the injury tolerance (~175g). However the upper neck tension was 250-350lbs. These values are in the range of the neck injury tolerance value of 254 lbs used in federal crash test standards. The 30mph crashes resulted in 70-75g head acceleration and 450-500lbs upper neck tension. The chest acceleration was 50-55g. The 40mph crashes produced 80-85g head acceleration and 500-700 lbs of tension in the upper neck.

I have investigated several real-world crashes that reflect the above results. Young children restrained in very severe frontal crashes. No evidence of head impact and no serious head injuries with severe tensile neck injuries (AO dislocations, spinal cord damage, etc).

Conclusions:

1. A severe crash (30mph) results in inertial loading of the head below head injury levels (assuming the occupant does not contact the vehicle interior). If a person is to shake (without impact) a 3-year-old child and cause serious head injuries then he/she has to produce loads more severe than a car hitting a wall at 30mph (30mph velocity stopping in less than 0.1 sec). This is absolutely impossible.

2. Even if someone could … During a situation where the torso is held and the head rotates as a result of inertial loading (non-impact), the neck forces exceed the injury tolerance before the head acceleration reached injurious levels. The 30mph crash resulted head accelerations below levels for serious head injury while still producing neck loads approximately double the accepted neck injury tolerance.

Notes: We did not measure angular acceleration of the head in these tests. Currently, similar 1-year-old crash test data is not available due to the fact that this age is usually tested in the appropriate restraint configuration using a rear-facing child seat.

-Michael Prange
At Chicago, Kirk T said at one point that sometimes the biomechanical engineer does not always know exactly what specific piece of data and analysis that the rest of us are looking for. At Chicago, Chris Van Ee said something to me indicating that in their MVA testings the real injurying forces on the neck are all basically Tension forces.

I've spent several days going over the the Prange 2007 article. I posted initial questions. Mike responded. I wrote up another set of questions for Mike OFF-LIST. Then I ask if we could meet with him and review the videos and the raw data. Then yesterday we were able to meet with Mike Prange at his office for 2 hours to go over the video and the raw data with my specific questions, trying to get at the specific points of analysis that I wanted clarified.

Attached in a fairly extensive word document (four pages) framing of the answers I think that I got, and how I think that these answers impact the legal world of SBS. I wrote this up this AM. Mike has not edited it. If I got things wrong, then they are my mistakes and I would appreciate correction, because I plan to use this in court and in making cases for appeals. [I've already given Prange 2007 along with Ouyang 2005, Luck 2006 to a lawyer in the middle of mounting an appeal.]

I've tried to be very clear in what I am thinking. However, if you are going to read my attached analysis, I might suggest that you have a copy of Prange 2003 and Prange 2007. I am sending a second e-mail with the Prange 2003 and Prange 2007 attached, if you don't have them handy.

Again I would appreciate feed back, even if it is negative, because I'm going with this for future testimonies and trials in the very near future.

John G

From my perspective, this is a major new article that counters several specific issues that we face:

- It is from 2007, and it involves PRIMARY research. In seeking a post conviction appeal and raising the issue of new evidence, this new evidence must meet the following conditions:
  1. The evidence has been discovered after trial and it could not have been obtained at or prior to trial through reasonable diligence;
  2. The evidence is not cumulative;
  3. It is not being used solely to impeach credibility;
  4. It would likely compel a different verdict.

This means we will always need new evidence, and this new evidence cannot be just a restatement of prior publications or rehashing of old data. It needs to be based on new primary data. When seeking an appeal, finding old articles that support our points does not appear to be helpful.

- Prange 2007 does demonstrate with primary new data that neck injury would be predicted BEFORE head injury in a MVA/impulse-loaded event, but we have been arguing this already, however, this new article is presenting Primary Raw Data supporting this.

- In my opinion, the raw data takes, “chin/chest” and “occiput/back” impact off the table. This is important.

THIS IS HUGE.

In consulting on cases, I have encountered several basic arguments from the proponents of SBS or accusing pediatricians to counter our claims that one should expect neck injury before head injury.

---First, the proponents of SBS argue that we don’t know the properties of real human neck. [We now have Duncan 1874, Ouyang 2005, and Luck 2006 to counter this.]
---Second, the proponents of SBS argue that it does not matter if the defense asserts that shaking alone cannot do it and that impact is required, because when the head/chin hits the chest and the head/occiput hits the back, you actually have impact. Therefore, everything that the defense is trying to say is ‘pure shaking’ is in reality, a ‘Shaken Impact’.
---Third, the proponents of SBS argue that the neck is not injured, because it is PROTECTED from injury by its very flexible and stretchable nature.
---Fourth, the proponents of SBS argue that the neck is not injured BECAUSE it is PROTECTED from stretch by the head/chin to chest and head/occiput to back impact.

This logic is demonstrated by the following quote from the testimony of a prominent proponent of SBS:

“neck injury certainly wouldn’t be characteristic of SBS basically because the child has such a large head that it is going to hit their chest and their back before they stretch their neck. And children also have very flexible necks flexible necks compared to adults. An infant won’t [have a neck injury] because there is no stretch on the neck. So, in essence, there’s not a really good reason to have a neck injury, but every once in a while. You rarely do see a case of that where it’s a ____ neck injury and not, like I said, the spinal fluid circulating around blood, Your could say, well, some of
that’s coming from the neck area, some of it’s in the back area, but that just means that’s incidental to SAH.”

When Dr. Prange shared this article, his summary stated, “During a 20 mph crash, the resulting head accelerations were 40-45 g, well below the injury tolerance (approx 175 g). However, the UPPER NECK TENSION was 250-350 lbs. These values are in the range of the neck injury tolerance value of 254 lbs used in federal crash tests standards. The 30 mph crashes resulted in 70-75 g head acceleration and 450-500 lbs upper neck tension. The 40 mph crashes produced 80-85 g head acceleration and 500-700 lbs of tension in the upper neck.

Conclusion:

- A severe crash (30 mph) results in inertial loading of the head below head injury levels (assuming the occupant does not contact the vehicle interior). If a person is to shake (WITHOUT IMPACT) a 3-year-old and cause serious head injuries then he/she has to produce loads more severe than a car hitting a wall at 30 mph (30 mph velocity stopping in less than 0.1 sec). This is impossible.

- Even if someone could …During a situation where the torso is held and the head rotates as a result of inertial loading (non-impact), the neck forces exceed the injury tolerance before the head acceleration reached injurious levels. The 30 mph crash resulted head accelerations below levels for serious head injury while still producing neck loads approximately double the accepted neck injury tolerance.”

He presented the “Peak UPPER Neck Tension” data in Figure 2.
He presented the “Peak resultant head accelerations” data in Figure 8.
He presented the HIC15 data in Figure 10, and HIC 36 data in Figure 11.

This seems quite clear and supports Prange’s above conclusions. However, it does not totally arm one to counter the proponents of SBS in court.

They will say that not only does the head hit the chest [and back] in a shaking, but also the head may sometimes even hit the knee. However, the head hitting the leg only occurred in the ‘belts-only’ testing, not in the Child Restraint System (CRS) testing. Abusive shaking with hands about the chest is most analogous to Car Restraint System (CRS) testing properly installed CRS with belt tension. CRS testing without belt tension has slack between CRS and engagement of deceleration forces and ‘belts-only’ testing had slack between infant and engagement of deceleration forces, and when deceleration forces did engage, they were not symmetrical and encouraged much bending at or below the waist. Also on viewing the video of the ‘belts-only’ testing, the hips are already flexed to >90 degrees BEFORE the start of the test, setting up the potential impact with the head. This configuration would never be present in an “abusive shaking”. As in the trial testimony quoted above, the proponents of SBS may claim that the head/chin to chest and/or head/occiput to back impact is the injuring impact, and that these impacts somehow protect the neck from injury.

Duhaime ’87 through Prange ’03 articles invite and suggest support for this challenge. I would suggest that you get out the Prange 2003 and go to Figure 2. Now look at the plotting of the “Angular Acceleration”. Note that the peak angular acceleration (2,600 rad/sec²) was recorded at
the EXACT MOMENT of impact of the head on chest or back and that this was very brief and that it was many times greater than angular accelerations anywhere else during the shaking cycle. This is a product of the ATD design with a single low hinge neck and the kind of ‘unnatural’ shaking motion that has to be generated in order to get this dummy to shake.

Note that when shaken, the newer ATDs like Carole Jenny efforts with the APRICA 3.4, a 7.5-pound dummy with upper and lower hinge, could only produce angular accelerations of 900-1400 rad/sec^2. On page 3 of 8 of the report, she states that, “Also the neck displacement is 155 degrees in the APRICA 3.4 and 95 degrees in the CRABI. The APRICA neck is much more flexible than the CRABI neck. The chin of the APRICA ATD gets closer to the chest and the back of the head goes further back.” (Carole Jenny’s report “Commonwealth v. Ann Power, December 29, 2005) NOTE, THAT SHE DID NOT SAY THAT THE HEAD ACTUALLY HIT THE CHEST OR BACK, BUT IMPLIED THAT WITH A BETTER NECK THAT IT MIGHT. I would also point out that she only got 900-1400 rad/sec^2 and basically confirmed Prange 2003, when they said that the 2600 rad/sec^2 was not the real number, but that it was the maximal potential that might ever be expected in a horizontal abusive shaking event.

THE UNADDRESSSED ISSUE IS STILL CHIN/CHEST AND OCCIPUT/ BACK IMPACT.
---Is it protective of the Neck?
---Is it in any shape or form of a magnitude that might be injurious?
To counter and shut down the SBS proponent’s arguments, we need raw data to prove that the answer to both of these questions is unequivocally “NO”!!

On the surface, the Prange 2007 article would appear to have not addressed these issues, and unless we can take these issues off the table, then the proponents of SBS will continue to hide behind these issues.

If they can be taken off the table, then every case should be subject to appeal, where the proponents of SBS have obtained a conviction and have asserted any of their counter arguments.

However, on close reading of the Prange 2007 article, particularly for the “CRS with Tension” data (the most analogous to abusive shaking dynamics), several things leap out:

First, the Peak Upper Neck Tension EXCEEDS Peak Lower Neck Tension at 20-30 mph crashes (compare Figure 2 with Figure 6).

Second, the Peak Upper Neck EXTENSION Moment was 2-4 times greater than the Peak Upper Neck Flexion Moment. Moment is TORQUE about an axis of rotation. The counter intuitive observation from this data is that the Peak FORCES responsible for generating rotational acceleration at the UPPER NECK occurred in EXTENSION—not FLEXION as when the head finally stops its forward motion.

If the raw data clearly demonstrates that in a real MVA with the forward facing child, the Peak Upper Neck Tension and Moment occur prior to the end of the forward head motion and that impact does not occur (or if it does occur that it is minimal and clearly benign) then these issues are off the table.
I met with and asked Dr. Michael Prange to review the raw data with me and explain it. I wanted to know exactly when in the forward arcing motion of the head that the “Peak Upper Neck Tension”/damaging stretching occurred. If it were prior to the end of the forward arcing motion, then Chin to Chest impact is **IRRELEVANT TO ANY CONSIDERATION OF NECK INJURY**. The proponents of SBS would be debunked in this area.

On the video and in the raw data, if one describes the arcing motion against the faced of a clock, starting from vertical (12:00) BEFORE the crash and ending at slightly past horizontal (3-3:30) at the end of forward motion; **Then the Peak Upper Neck Tension/damaging stretch occurred between 1:00-1:30. IT IS NOT RELATED TO IMPACT AT ALL. By this data, the potential for chin/chest impact or the flexibility of the neck cannot be asserted to be in anyway protective of the neck.**

The second issue was with peak angular acceleration. Torque about an axis is what generates rotational acceleration. The Prange 2007 (Figure 3 vs. Figure 4) indicates that the EXTENSION MOMENT/torque was 2-4 times greater than the Flexion Moment/torque. Although they did not actually record angular acceleration, the peak angular acceleration of the head would be expected to occur when the peak torque was acting. We needed to know when this was occurring and did it in anyway correspond to the final forward motion of the head and/or impact of the head/chin on the chest.

On the video and in the raw data, it is clear that the answer is ‘NO’. The Peak Upper Neck Extension Moment occurred approximately between 1:00 and 2:00 on the clock face—not at the end of forward motion (3:00-3:30) and certainly not corresponding to any impact of head/chin on chest.

In my opinion, the raw data behind the Prange 2007 article clearly takes head/chin to chest impact (and head/occiput to back impact) off the table. It is not protective of the neck, and it is not a contributor to Peak Angular Acceleration---even if the neck were flexible enough to allow it to occur.

In hindsight, this is obvious. SBS was originally described as ‘whiplash’ injury. Whiplash occurs when something (head) is moving in one direction by inertia (momentum/velocity x mass, but not accelerating) and is then has its motion jerked/stopped and accelerated in the opposite direction by a force. As the head is moving forward at first with the body and the body is then accelerated in the opposite direction, the head initially continues forward by inertia, until the neck suddenly becomes taut and JERKS the head. This is the point of maximum tension stress on the upper neck and by the raw data the point of maximum torque that would be causing peak angular acceleration of the head.

The Prange 2007 article and the raw data behind it should make every prior SBS/Shaken Impact conviction open for appeal on the basis of NEW EVIDENCE, if there was testimony during the prosecution’s case-in-chief that Head/Chin to chest or Head/Occiput to back impact constituted the “Impact” part of Shaken Impact Syndrome or that these hypothesized impacts were somehow protective of the neck from stretch.
From: Infant injury evaluation [EBMS@PEACH.EASE.LSOFT.COM] on behalf of Kirk Thibault [kthibault@BIOMECHANICSINC.COM]
Sent: Thursday, June 28, 2007 1:28 PM
To: EBMS@PEACH.EASE.LSOFT.COM
Subject: Re: Prange 2007---critique and raw data

Attachments: composite.jpg

John - on the very specific topic of the timing of peak neck tension and head impact (i.e., bottoming out of the head/chin on some structure).

Attached is a composite of data taken from a sled test simulating a 35 mph frontal collision. The data are from the Hybrid III 3 yo that was forward facing and restrained in an appropriate 5point harness. The plot overlays the resultant head acceleration and the neck force in the axial (Fz) direction as a function of time. I have not seen Michael's raw data (accel or force v time curves) but I suspect they are similar.

The notable point is the peaking of the tensile neck loads prior to the head striking anything (i.e., neck forces peak due to the inertial loads acting on the head neck system while the torso is restrained and the system is undergoing frontal decel). It is interesting to note that the inertial head accel peak coincides with the peak in the Fz, as one would expect. There is a spike in the head accel trace following this peak due to the head striking the lap of the dummy.

Is this what you are interested in understanding? Your argument, although interesting, is somewhat academic, as the loads experienced in this type of event are substantially greater than those experienced during shaking. I have not heard the "neck being protected by the chin bottoming out on the chest" yet, and, to my knowledge, the only person who really makes a big deal about this contact is Cory, arguing that it substantially increases the head accelerations (Cory and Jones reported peak resultant head accels with their shaking doll of over 100g and HICs > 1000 because of the spikey nature of this contact - I'm not sure how legit their model was in this regard).

Kirk
35 mph frontal sled test
3yo HIII, restrained in forward facing, 5-point restraint.

*spike in head accel due to head impact with lap.
Kirk,

This is exactly the kind of data I was looking for.

Mike's data was collected from an ATD with 6 way sensors at the upper and lower neck--so it gave a graph of upper neck tension and lower neck tension vs time. It also had a graph of upper neck moment and lower neck moment vs. time. None of his CRS tests resulted in head impact with the lap, legs, or knees. The peak upper neck tensions occurred at or very near the peak upper neck EXTENSION moment, and both of these occurred well before end of the arcing motion. This was specifically what I was looking for. Your peak total neck tension (not broken into upper and lower) and peak head acceleration (if you ignore the head lap contact), which I think corresponds to Mike's peak upper neck EXTENSION moment--looks similar to his. Again his data divided upper neck moment from lower neck moment and extension moment from flexion moment.

I appreciate that these numbers are for 30+ mile MVA's and that the forces and accelerations that would have to be exerted on the chest of infant or toddler or 3 year old to reproduce these accelerations are many times beyond human strength. However, armed with this I feel very good about saying that neck injury is clearly predicted before head injury (that is not new), BUT now when they come back with an explanation for why the neck is not injured that includes anything like:
---we don't know the strength of an infant's neck;
---the increased flexibility and infantile anatomy somehow protects the neck; or
---head impacting on the chest and/or back is damaging impact of a 'pure shaking' event that makes it, in fact, a "Shaken Impact", we can now specifically confront them with this recently published data. We can actually try to draw them into saying or speculating on these issues, and then try to back them into an impeachable position. The jury won't understand the physics but if we can make them squirm, stutter, or confess that they are totally unaware of new literature --that should to some degree help our case.

Again, in my opinion, any case that resulted in a conviction where the SBS proponent made these kinds of statements to discredit the defense's claim that neck injury should occur before head injury in cases of impulse loading, then in my opinion we now have legitimate grounds to say that new evidence has recently come forward to prove them wrong and that this is new evidence and therefore warrents a new trial.

New evidence is apparently tricky. I worked on an appeal case where trial was third week of November and the article I would like to have claimed as new evidence came out the first week of November. I could not claim it was new.

The quote from a prior testimony of a proponent of SBS that I included is real; the lawyer in an upcoming case where this expert is expected to testify for the prosecution just sent it to me.

John G
From: Infant injury evaluation [EBMS@PEACH.EASE.LSOFT.COM] on behalf of jgalaznik1@AOL.COM
Sent: Friday, June 29, 2007 8:54 AM
To: EBMS@PEACH.EASE.LSOFT.COM
Subject: Re: Prange 2007---critique and raw data

Kirk,

I've had a chance to sleep on your response.

"Cory and Jones reported peak resultant head accels with their shaking doll of over 100g and HCl>1000 because of the spikey nature of this contact--I'm not sure how legit their model was in this regard."

This sounds like the kind of report out of Cory and Jones that would lead the pediatric proponents of SBS to make statements in court like those I have reported. If the head were to have to be stopped by the forces coming through the neck, 100Gs of head accel and a 2-3 pound head would require 2-300 pounds of force to stop it and even they knew this would break the neck. They could not allow this data to come out without offering an explanation to their followers that might be used in court to preserve the possibility that pure shaking can still do it. They want the 100Gs and the HIC >1000 for brain damage to make the case that "shaking alone" can to it. But then they needed an explanation as to why the neck did not break in order to get head injury BEFORE neck injury.

ANSWER: The neck is protected from stretch by this damaging IMPACT of the chin with the chest and occiput with back.

This "answer" coupled with the rumored report of 100Gs and a HIC> 1000, is all a pediatrician needs to confidently make the arguments in court that I mentioned---1. real properties of neck not known. 2. the impact of a pure shaking is of such a magnitude as to make every pure shaking in affect a "Shaken Impact". 3. The neck is super flexible and stretchible and easily allows this impact. 4. The neck itself is 'protected' from damaging forces in this 'violent abusive shaking' by the forces being absorbed by the chin/chest and occiput/back impact. [That actually sounds logical enough to a jury to buy in to--how can you blame them for not believeing it?]

What did Cory have to do to get to 100Gs ---even if only a brief spike----???use a single low hinge neck and a "gravity assisted" motion (vertical not horizontal) starting with the ATD above shoulder height and ending with the ATD at knee/distal thigh level and have a steel plate on the chin and another on the chest????

Mike's data didn't appear to have a 100G peak, occurring at the end of the forward arcing motion that I would think should correspond to impact of chin on chest. If the chin is making contact in Mike's data--Bottoming Out on the Chest--then it would have appeared to have done so 'gently' without any measured impact spike. To me, this is what Challenges/Debunks their claim that this is the damaging head impact--the 100Gs and HIC>1000.

And since the damaging Peak Upper Neck Tension had already occurred back at the 1:00-1:30 head position in the forward arcing motion--then this gentle bottoming out/impact of chin/chest does not protect the upper neck from damage.

----------

Mike's data did contain "Peak Lower Neck Flexion Moment" and it was substantial (His Figure 7), but peak lower neck tension were < upper neck tension at 20-30mph velocity crashes and = upper neck tension at 40 mph crashes. I think that he also indicated that there was no compressing forces
measurable in the anterior neck---leading him to indicate that the neck acts like a rope. This makes me wonder if the ATDs chin is "gently" bottoming out on the upper chest/anterior base of the neck BEFORE the final terminal stopping of the heads forward arcing motion. If chin were to actually have the rigidity to serve as fulcrum, then:
---if base of neck to CG of head were 8cm
---if tip of chin to CG of the head were 8cm
---if bottomed out chin/chest point were 3-4 cm from the lower neck sensor.

Then WHAT?

All forces at the lower/base of the neck would be tension---Mike would still say that the neck is acting like a rope. The Flexion moment at the upper neck would be small---and it has bottomed out and moment at the upper neck minimized. The axis of rotation would now be about the point of chin/chest contact and the lower neck would to some extent be protected because pure torque would have been tranformed to tension. This would result in some protection of the lower neck in an MVA and predominance of upper neck injury when neck injury is observed.

Again I would appreciate feedback.

John G
John - your emails are so packed with things that I'll have to take some time to digest and respond. In the meantime, attached is Table 3 from Cory and Jones, 2003 where they report the peak tangential accelerations of the head (among other parameters) and HIC for various tests. As the numbers for acceleration are given in units of meters/second$^2$, just divide by 10 to get a quick estimate of acceleration in units of "g" ($g = 9.81 \text{ m/sec}^2$ - close enough).

This is where I was getting those figures - it is not "rumor", it is published data. The pulse width they report for these peaks is on the order of 15-20 milliseconds. Obviously this gets the head accels and HICs nice and high to make the point that shaking can cause accels much higher than those reported in the 1987 Duhaime, et al study. There is no mention of the neck in this paper and the effect that this contact has on the neck loads. I suppose whoever is arguing that the contact between chin and chest is protective is basically arguing that this contact prevents the neck from exceeding some non-injurious range of motion with respect to angular displacement. However, as we have already pointed out, the neck tension is important. So, when you look at Duncan's tests, for example, he applied axial tension to the surrogates he tests. Presumably the neck angular range of motion was not exceeded, but he managed to cause structural disruption of the spine due to the axial forces applied in tension.

Cory and Jones do not publish the acceleration-time curves for their tests so it is not possible to compare their shaking (with head contact to the torso) events to anything. They only publish peak values and the width of the "peak" event, whatever that means. There is no clip value specified, so it is not possible to determine where or how they are expressing the width of the "spike". This is sort of a classic case of one trying to have their cake and eat it too. We want the head accels to be real high, generated by shaking with impact, but we want the neck loads to be real low, from a protective impact no less. So where are the mandible injuries, bruising to the chest, etc.? They are in the same pace as the bruising and grip marks on the torso that never seem to be present, despite underlying posterior rib fractures.

Kirk
Table III. Results recorded from the body and head accelerometer data and the calculated Head Injury Criterion (HIC) values for the parameter combination model shake tests.

<table>
<thead>
<tr>
<th>Test-Name</th>
<th>Peak tangential head AccN ($m/s^2$)</th>
<th>Peak angular head AccN ($rad/s^2$)</th>
<th>Peak tangential head velocity (m/s)</th>
<th>Peak angular head velocity (rad/s)</th>
<th>Time duration of peak head AccN curve (ms)</th>
<th>Head Injury Criterion (HIC)</th>
<th>Peak body AccN ($m/s^2$)</th>
<th>Peak body velocity (m/s)</th>
<th>Time duration of peak body AccN curve (ms)</th>
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<td>8149.80</td>
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I've misplaced my copy of Cory 2003, but if memory serves me right---this was a single low neck hinged ATD of about 10 pounds (they also shook a heavier ATD and got lower numbers--didn't want that so the abandoned that). The Tangential velolcity and acceleration was measured and the angular head acceleration was calculated--not measured. I thought it was similar to the Duhaime '87 and the Prange 2003 ATD. [Prange 2003 was single hinge--no resistance--no limit to ROM except by impact with body--and 9.2cm below the CG of the head] I also thought that in the plain old fashioned horizontal "classical shake" (although it would have to be with the unnatural motion inorder to get it to shake) that their numbers were similar to Duhaime '87.

This model does not dissipate energy to get the head up and over the shoulder and neck because the shaker has to sneak some rotation or up-and-down motion into the act. Actually, this design of ATD shakes best with an up-and-down motion rather than a horizontal motion. The up-and-down motion sends the head flying just like it is supposed to. This model does not dissipate energy at the "jerk" of peak upper neck tension at 1-1:30 point in the arcing motion as in the CRS srash tests. It does send the head crashing into the chest and this impact is the total stopper of the head's tangential velocity--Like the impact spike in the Prange 2003 Figure 2 that got the 2,600 rad/sec2 and I presume for Duhaime '87.

Is the Cory Table III that you sent for their "classic shake" or for their "GRAVITY ASSISTED" shake?? If the Proponents of SBS are quoting numbers from Cory's GRAVITY ASSISTED (vertical shake); and making and showing videos of a "classic horizontal" shake--this in and of itself is intentional (criminal) fraud.

In looking at their table, I was puzzled. If I want to know the value of their Radius of the circle of their rotation---I should be able to calculate this by dividing the Peak angular head velocity (rad/sec) into the Peak tangential head velocity (Meters/sec) and get Meters/Radian.

8.66meters/sec  /  50.95Rad/sec = .16meters/radian or each radian is 16cm. If CG of the head is 8-9cm above the base of the neck, where is the center of their ATD's rotation---mid thorax?? or is shakers wrist that is the rotational center?

John G