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Can Shaking Alone Cause Fatal Brain Injury?

A biomechanical assessment of the Duhaime shaken baby syndrome model

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ABSTRACT

A biomechanical model of a one-month old baby was designed and tested by Duhaime and co-workers in 1987 in an attempt to assess the biomechanics of the shaken baby syndrome (SBS). The study implied that pure shaking alone cannot cause fatal head injuries, a factor which has been applied in criminal courts. In an attempt to test the validity of the model a preliminary study was undertaken in which a replica was constructed and tested. The broad description of the design and construction of the Duhaime model allowed for variations and therefore uncertainties in its reproduction. It was postulated therefore that differences in certain parameters may increase angular head accelerations. To further investigate this observation, an adjustable replica model was developed and tested.

The results indicated that certain parameter changes in the model did in fact lead to an increase in angular head acceleration. When these parameter changes were combined and an injurious shake pattern was employed, using maximum physical effort, the angular head acceleration results exceeded the original Duhaime et al. (1987) results and spanned two scaled tolerance limits for concussion. Additionally, literature suggests that the tolerance limits used to assess the shaking simulation results in the original study may not be reliable. Results from our study were closer to the internal head injury, subdural haematoma, tolerance limits. A series of end point impacts were identified in the shake cycles, therefore, an impact-based head injury measure (Head Injury Criterion – HIC) was utilized to assess their severity. Seven out of ten tests conducted resulted in HIC values exceeding the tolerance limits (critical load value, Stürtz, 1980) suggested for children.

At this present stage the authors conclude that it cannot be categorically stated, from a biomechanical perspective, that pure shaking cannot cause fatal head injuries in an infant. Parameters identified in this study require further investigation to assess the accuracy of simulation and increase the biofidelity of the models before further conclusions can be drawn. There must now be sufficient doubt in the reliability of the Duhaime et al. (1987) biomechanical study to warrant the exclusion of such testimony in cases of suspected shaken baby syndrome.

INTRODUCTION

A clinical, pathological, and biomechanical study of the shaken baby syndrome (SBS) was conducted by Duhaime et al. in 1987. The biomechanical study required volunteers to 'violently' shake an infant model. The angular head acceleration output values did not exceed scaled tolerance limits for fatal infant brain injury, thus suggesting that pure shaking by a human cannot cause fatal brain injury in young infants. However, other clinical and pathological studies provide evidence to the contrary, suggesting that pure shaking (without an associated impact) can cause death in young infants (Hadley et al., 1989; Alexander et al., 1990; Gilliland and Folberg, 1996).

The biomechanical response characteristics of young infants were not investigated or modelled in the Duhaime et al. (1987) study. Although infant response data is sparse, there are some guidelines (scaled from adults) available from biomechanical studies within the automotive industry. Preliminary experiments conducted by the authors suggest that slight changes in the mechanical parameters of the Duhaime model might affect the angular head acceleration values obtained during shaking experiments. This paper documents a re-investigation of the Duhaime et al. (1987) biomechanical model.

BACKGROUND

Guthkelch first described the mechanism of injury now known as the shaken baby syndrome in 1971. In 1972 Caffey described the shaking of infants and suggested as part of his conclusions that 'there are several features of infantile subdural haematomas which indicate that they are not usually caused by direct impact injuries to the head, but are caused by indirect acceleration-deceleration traction stresses such as whiplash-shaking of the head. These include bilaterality of subdural haematomas in 85% of infants (Ingraham and Matson, 1954) and frequent bilateral retinal haemorrhages. There is a striking lack of signs of impact injuries such as blows to the head. Usually there are no bruises to the face or scalp, no subperiosteal cephal-haematomas, and no fractures of the calvaria' (*sic*).

Duhaime et al. (1987) suggested that a history of shaking was usually lacking in so called 'SBS cases'. They reviewed all cases of SBS at the Children's Hospital of Philadelphia between January 1978, and March 1985. Thirteen of 48 suspected shake injuries resulted in fatalities, all had evidence of blunt head trauma at autopsy. The authors commented of their study that 'it is of interest that in more than half of our fatal cases, no evidence of external trauma was noted on the initial physical examination, which helped to contribute to the diagnosis of "shaken baby syndrome".'

The biomechanical section of the study consisted of a model of a one-month-old infant (with three different neck sections and two head types) instrumented with an accelerometer transducer to measure peak tangential head acceleration. Volunteers were asked to violently shake and subsequently to impact the rear of the head (occiput) of the model against a metal bar and a padded surface. The head acceleration output values were compared with an acceleration tolerance curve, scaled from animal experiments to the brain mass of an infant (500 gm). Duhaime et al. (1987) concluded that 'the angular acceleration and velocity associated with shaking occurs well below the injury range, while the values for impacts span concussion, subdural [haematoma], and diffuse axonal injury ranges. This was true for all neck conditions with and without skulls'. 'It is our conclusion that the shaken baby syndrome, at least in its most severe acute form, is not usually caused by shaking alone.'

Duhaime et al. (1987) suggest that 'although shaking may, in fact, be part of the process, it is more likely that such infants suffer blunt impact' (that is, a shake and then an impact against a crib or other surface).

Literature on the shaken baby syndrome often fails to distinguish between 'pure shaking' and shaken impact baby syndrome. Therefore, for the purposes of this paper the SBS is defined as the manual shaking of an infant *with or without* associated head impact. Pure shaking describes the shaking of an infant *without* an impact of the head to an object, i.e. not excluding the impact of the chin-to-chest or occiput-to-back of the infant.

Bruce and Zimmerman (1989) summarised the biomechanical findings of Duhaime et al. (1987) and supported their conclusion, commenting that 'the physician should be suspicious of a report of pure shaking, as this is an unlikely scenario. If criminal prosecution is to occur, the charges should include impact injury as well as shaking'.

Hadley et al. (1989) commented on Duhaime et al's (1987) findings and conducted a study to investigate the problem further. All cases of non-accidental head injuries in infants were examined, with particular attention given to those with neurological symptoms, but no evidence of direct cranial trauma. All the patients defined as being in the 'isolated whiplash-shake injury subgroup' were very young infants (14 months or under). Hadley et al. (1989) pointed out that 'five of six patients on whom an autopsy was performed had no post-mortem evidence of direct cranial trauma'. The authors suggested that 'the discrepancies between this review and that... by Duhaime et al. (1987) may be explained in part by the younger age of the patients in our series (median age, three months; range 1.5 to 14 months) versus theirs (median age, ten months; range three to 24 months). The high incidence of subdural haematoma in our 13 whiplash-shake injury patients and the high incidence of cervical cord injury in the six patients in whom autopsy was performed presumably also relates to the very young, immature patients in the population in our study'.

Hadley et al. (1989) suggest a note of caution in the argument for shaken-impact commenting that 'direct cranial trauma is not always required. We believe that a subgroup of patients, particularly the very young, may sustain severe neurological morbidity and mortality from rapid acceleration-deceleration injuries incurred from a whiplash-shake mechanism alone'.

Gilliland and Folberg (1996) reviewed a number of studies suggesting 'as more reports of systematic and ocular findings at death have been described it has become evident that *many* of the babies believed to have been shaken have suffered impact injuries (Duhaime et al., 1987; Rao et al., 1988; Hadley et al., 1989; Alexander et al., 1990; Elner et al, 1990, Massicotte et al, 1991, Munger et al., 1993; Bundez et al., 1994). To investigate whether shaking without direct head trauma is sufficient to inflict a lethal injury in infants Gilliland and Folberg (1996) reviewed findings from a large series of child deaths. The previously referenced studies indicated that 'many' and therefore not all 'of the babies believed to have been shaken have suffered impact injuries'. Their results (Gilliland and Folberg, 1996) showed that 'nine (11.3%) of the 80 head-injury deaths met the definition of death by the exclusive shaking mechanism of injury'. They suggested that their study 'confirmed earlier observations that *some* shaken babies do not have evidence of blunt head injuries (Rao et al., 1988; Hadley et al., 1989; Alexander et al., 1990; Massicotte et al., 1991; Munger et al., 1993; Bundez et al., 1994)'.

Another study by Jacobi (1986) was discussed where three infants (mean age six-months) were fatally injured from shaking alone (defined clinically). Gilliland and Folberg (1996) suggested that 'shaking alone was not as often fatal as direct impacts but it was a lethal mechanism of injury in the child abuse deaths in children in Jacobi's study (1986)'.

Gilliland and Folberg (1996) criticise the biofidelity (i.e. human-like characteristics) of the Duhaime biomechanical model and suggest that 'the Duhaime et al. (1987) model is inadequate to explain death in children with no scalp and skull injuries after complete autopsy examination'.

Alexander et al. (1990) found that five out of nine of the fatally injured infants in their study had no evidence of impact at autopsy. The authors point out that 'the argument has been made that children with intracranial injuries but without detectable signs of external head trauma may have suffered an impact with a padded surface, such as a cushion or crib mattress, and this impact caused an intracranial injury. This cannot be completely refuted'... 'providing the burden of proof would seem at this point to fall to those who claim that impact must be present in all instances of serious intracranial injury'.

The results from the Duhaime biomechanical simulation are used to suggest that pure shaking has been shown to produce insufficient force to cause fatal head injury in infants. These results are sometimes used in alleged shaken baby syndrome cases to defend an assailant's actions, suggesting that even if shaking did occur, shaking alone cannot cause death in infants. However, Duhaime et al. (1987) did not categorically state this, they suggested that 'the shaken baby syndrome, at least in its most severe acute form, is *not usually* caused by shaking alone'. In 1996, Duhaime further attempted to clarify the conclusions drawn from the 1987 biomechanical simulation, stating that the researchers 'never said that beyond a shadow of a doubt shaking cannot cause injuries'. What they said was that 'shaking, at least with this model, produces angular decelerations which are too small to cause the target injuries for which there are established thresholds'. Again in 1998 Duhaime et al. suggested that 'although there is considerable controversy, the available evidence suggests that it is the sudden deceleration associated with the forceful shaking of the head against a surface that is responsible *for most, if not all*, severe, inflicted brain injuries'. Again there is no categoric statement that pure shaking cannot ever produce fatal head injuries.

EXPERIMENTAL METHODOLOGY

1. Preliminary Investigations

A preliminary study was conducted which aimed to replicate the Duhaime model and allow direct observations of the biomechanical study, highlighting any problems associated with the construction and testing of the model. Analyses were also conducted on the reproducibility and repeatability of the experiments.

A second preliminary study involved a kinematic analysis of variation in the individual shaking styles/patterns. The study permitted an analysis of the effects of an increase in mass (corresponding to an increase in age of the infants) on volunteers' input accelerations to the model body. The results from both studies assisted in the formulation of a methodology for the main study.

During the construction, testing and analysis of the replica Duhaime model it was observed that volunteers demonstrated more varied shake patterns than just the standard anteroposterior (A-P) shake, suggested by Duhaime et al. (1987).

Head acceleration values derived from the replica study were higher than those recorded by Duhaime et al. (1987). In addition, some of the variant shake patterns demonstrated higher head accelerations than the A-P shake pattern.

To fully investigate the variation in volunteer shaking patterns, found during the preliminary study, a kinematic analysis was conducted using additional volunteers. Three models were constructed corresponding to the respective mean mass values of a one month-old, seven month-old and 18 month-old child. Eleven volunteers were requested to shake the model with maximum physical effort but no instruction was given as to the shake style or pattern to be adopted during testing. The study illustrated that individuals who receive no instruction as to how to shake the model adopt very different patterns and in fact do not always adopt the anteroposterior shake pattern suggested by Duhaime et al. (1987). The accelerations achieved by all volunteers were reduced as the mass of the model (corresponding to a greater infant age) was increased. This suggested that the findings of Duhaime et al. (1987) should be interpreted only after the consideration of the mass of each individual child. For example, if an infant weighs 10 kg the findings of Duhaime et al. (1987) cannot be directly applied, since the level of input and body acceleration during shaking is likely to be reduced due to the greater mass of the infant. It is therefore imperative that research findings are applied appropriately in any given case.

It is noteworthy that during the shaking tests all seven female volunteers and some male volunteers showed signs of severe fatigue after ten seconds of violent shaking and reported that they would have found it difficult to continue for much longer. Therefore, further work may prove that the shake durations postulated to cause brain injury, sometimes suggested to be as long as one to two minutes, might be physically difficult to achieve.

In addition to the potential variability in acceleration values introduced by different shake patterns, it became apparent, during the production and testing of the Duhaime replica model, that there was considerable potential for variation in the way the model could be constructed from the vague description provided in the Duhaime et al. (1987) paper. It was suspected that the biomechanical characteristics of the neck, centre of gravity, chest and back padding and neck insertion point of the model might influence the level of peak angular head acceleration produced.

To assess the relative significance of each of the biomechanical parameters, an adjustable model was designed and constructed which allowed these pre-defined parameters to be changed from a standard model. Once the model was constructed a parametric study was conducted to ascertain the effect of changes to the model on the peak angular head acceleration, and therefore head injury level, during shaking tests.

2. Main study

The aim of the main study was to investigate:

> *'What factors significantly affect the angular head acceleration of a biomechanical model constructed to simulate the shaken baby syndrome?'*

The adjustable model was designed to fulfil the following design specifications:

- The standard model must be a replica of the Duhaime et al. (1987) model.
- The model must be designed to allow replacement of the standard head with a modified head with a more realistic neck insertion point.
- The model must facilitate the changing of neck type from thin rubber neck to thick rubber neck and metal hinge neck.
- The model must allow for the adjustment of the mass attached to the thorax thus moving the centre of gravity of the body.
- The model must facilitate the changing of chest and back material.

Head

Standard head

The design and construction of a standard head followed the methods documented by Duhaime et al. (1987). In addition, it incorporated a metal collar to facilitate the attachment of the head to various neck configurations. The head-neck collar was placed in the desired position and resin (Fastcast, polyurethane UREOL 5202-1 A/B plus Filler DT 082, supplied by John Burn & Co. Ltd. (Birmingham, UK) was mixed and poured around the collar to secure it within the neck cavity of the model head. The final head mass (including head/neck collar) was 830 gm which fell within the range (770–870 gm) suggested by Duhaime et al. (1987). Head dimensions approximated to those of the Duhaime model.

Modified head

A modified head was constructed using the same methodology as the standard head. However, the head-neck collar was attached further inside the head than with the standard head at an accurate insertion point i.e. at the 0C junction (occipital condyles). The insertion position was estimated by scaling to the dimensions of a one-month-old infant's head using dimensions from anthropometric databases (Snyder et al., 1977; Norris and Wilson, 1995) and the actual dimensions of the model head using a method documented in Irwin and Mertz (1997) on the CRABI dummy (Child Restraint Air Bag Interaction). The head was cut to the required size (Cory, 2001) and shape to allow the head-neck collar to be placed in the required position. Resin was again used to hold the collar in place and to seal the remaining open parts of the head.

Necks

Three neck types were constructed to replicate those described by Duhaime et al. (1987), two rubber tube types and a metal hinge neck.

Rubber necks

The material used in the original study was 'commercially available hollow red rubber tubing' (Duhaime, 1998). The material used in the adjustable model was 'tubing-red rubber' supplied by Philip Harris-Scientific (Cardiff, U.K.). Neck dimensions approximated to those of the Duhaime model. As with the original model the thin rubber neck did not support the weight of the head in the upright position but did not kink when the head was allowed to fall unsupported. The thick rubber neck was able to support the weight of the head in the vertical position but allowed full passive movement of the head.

Both necks were cut to 4 cm (Duhaime et al., 1987) and mild steel tube plugs were bonded (using Araldite rapid adhesive) into ends of each neck to facilitate the attachment of the head to the neck and the neck to the thorax of the model. Once the adhesive had set, holes were drilled though each side of the neck and plug and screws were used to improve the strength of the join.

Metal hinge neck

The metal hinge neck was a replica of that used in the original study. It was a '360-degree steel hinge, 3.6 cm in width, placed in a horizontal plane to allow complete anteropos-

Table I. Description of each model configuration for the shake tests.

Model name	Neck type	Position of centre of gravity from the head	Chest and back padding	Head type	Shake type
Standard model	Thin rubber neck (medium resistance)	24 cm	Silicone	Standard head	Standard anteroposterior – low physical effort
Thick rubber neck model	Thick rubber neck (high resistance)	24 cm	Silicone	Standard head	Standard anteroposterior – low physical effort
Hinge neck model	Hinge neck (low resistance)	24 cm	Silicone	Standard head	Standard anteroposterior – low physical effort
High centre of gravity model	Thin rubber neck (medium resistance)	22 cm	Silicone	Standard head	Standard anteroposterior – low physical effort
Cotton wool chest and back model	Thin rubber neck (medium resistance)	24 cm	Cotton wool	Standard head	Standard anteroposterior – low physical effort
Modified head model	Thin rubber neck (medium resistance)	24 cm	Silicone	Modified head	Standard anteroposterior – low physical effort
Parameter combination model	Hinge neck (low resistance)	22 cm	Cotton wool	Standard head	Gravity assisted– maximum physical effort

terior angulation of the head. The centre of rotation was 3.3 cm below the estimated level of the skull base (approximating the C-6 vertebral level)' (Duhaime et al., 1987). The total hinge neck length was 4 cm. A nylon sleeve was added to the inside of the hinge to reduce the effects of friction.

Body

Frame

The design of the frame facilitated the attachment of body masses, different neck types, shoulder and chest and back pouches. Anthropometric studies were consulted for approximate dimensions for shoulder width and shoulder to buttock height for the one-month-old age group (Snyder et al., 1977; Steenbekkers, 1993; Pheasant, 1988; Beusenberg et al., 1993).

A T-bar design was utilised with an upper horizontal (shoulder) section and a midline vertical (vertebral) section. Body masses were constructed to add to the T-bar to ensure total body mass of 3-4 kg (Duhaime et al., 1987) and to enable the centre of gravity of the model to be adjusted from a more central position (standard model mass configuration) to a higher centre of gravity.

Shoulder, chest and back pouches

A similar shape and material (cotton) to the Duhaime model was used to design and construct the body section for the adjustable model.

The two types of chest and back padding were cotton wool and silicone. The cotton wool was 50% cotton and 50% viscose and the silicone was a two-part system (Burnco Silicone rubber with catalyst type F (fast), supplied by John Burn & Co. Ltd. Birmingham, U.K.).

Details of the model configurations can be found in Table I. More detailed information on the design and construction of the model can be found in Cory (2001).

Instrumentation and data capture

An accelerometer (Bruel & Kajaer type 4369) was attached to the thorax of the adjustable model (sampling at 5000 Hz) to record acceleration and thus allow the subsequent calculation of velocity. In accordance with the Duhaime model, tangential head acceleration was recorded utilising a piezoelectric accelerometer (PCB Piezotronics Model no. 339B10) attached to the vertex in a coronal plane through the centre of the neck. *Acknowledge*

(BIOPAC Systems, Inc., CA 93117, USA) software was utilised for analysing the accelerometer data.

Test Procedures

Shake type and holding pattern

In the first series of tests (parametric tests) the volunteer was asked to hold the model at a position equivalent to under the arms of an infant and shake it in an anteroposterior direction keeping the shake pattern the same throughout all tests. The volunteer adopted a low acceleration shake type to allow multiple consecutive controlled tests and to maintain the shake characteristics as uniform and constant as possible. All test procedures were the same for each parameter change on the adjustable model.

In a second series of tests the model was configured such that the parameters which had produced the greatest accelerations individually were combined and a model constructed which would produce the greatest acceleration as a whole (the worst case scenario). In addition, the volunteer was asked to use a shake pattern, which had been shown to produce the greatest level of acceleration, the 'gravity assisted' shake pattern. In the 'gravity assisted' shake pattern the arms are extended such that the model is elevated above one shoulder and accelerated downwards to below waist level (using gravity to assist). This results in the back of the head (occiput) impacting with the back of the model. The volunteer's arms are then pulled upwards, returning to the original position above the shoulders (with the volunteer's head tilted to avoid collision with the model) and inwards to induce chin-chest impact at the opposite end-point of the shake cycle.

RESULTS

A series of shaking tests were conducted to determine significant biomechanical parameters on an adjustable model of an infant. Although the volunteer was instructed to shake the model in the same manner for each test, some variation from test to test was expected. Therefore, the variation between tests in peak body acceleration (i.e. the input acceleration imparted to the body of the model by the volunteer) may affect the peak angular head (output) acceleration (i.e. acceleration of the head of the infant model). A rationalising method was employed to account for the slight increase/decrease in body acceleration. The ratio of the two values was calculated [peak head acceleration (rad/s^2)/ peak body acceleration (m/s^2)]. This 'ratio' value resolves the differences being attributed to a change in input acceleration and not parameter change. Therefore, the 'ratio' value was utilised to assess the statistical significance of parameter changes to the model.

Ratio

The Levene's test was used to test for differences in variance between the standard model ratio results and ratio values for models with each subsequent parameter change. The appropriate Student's t-Test (that is, for equal or unequal variance) was then used to test for significant differences between the standard model ratios and each set of ratio results for every parameter change. Table II shows the ratio value and the t-Test results for each parameter change.

Figure 1 shows results for the parametric study plotted in the ratio format, that is, angular head acceleration (rad/s^2) against linear body acceleration (m/s^2).

Adjusting all significant parameter changes-parameter combination model

It can be seen from the results presented in Table II that the hinge neck, high centre of gravity and the cotton wool chest and back models showed significant differences to the 'ratio' values, from the standard model configuration. Therefore, as documented in Table I the parameter combination model included the hinge neck, high centre of gravity and cotton wool chest and back. The shaking type was changed to 'gravity assisted' with maximum physical effort from the shaker.

Figure 2 shows results for the parameter combination model plotted in the format used by Duhaime et al. (1987), that is, angular head acceleration (rad/s^2) against angular head

Table II. Statistical results for ratio values of standard model tests compared to ratio values for each adjusted parameter test.

Model type	Mean (ratio)	Standard deviation	Variance	T-test used	T-test result (P value) when compared to standard model	Significant difference to standard model (P<0.05)
Standard model	31.01	4.30	18.51	–	–	–
Thick rubber neck model	32.50	3.02	9.13	Assuming unequal variances	0.128	No
Hinge neck model	72.09	13.44	180.55	Assuming unequal variances	0.000	Yes
Higher centre of gravity neck model	34.18	3.44	11.80	Assuming unequal variances	0.003	Yes
Cotton wool chest and back model	42.54	4.27	18.23	Assuming equal variances	0.000	Yes
Modified head model	32.72	4.12	16.97	Assuming equal variances	0.121	No

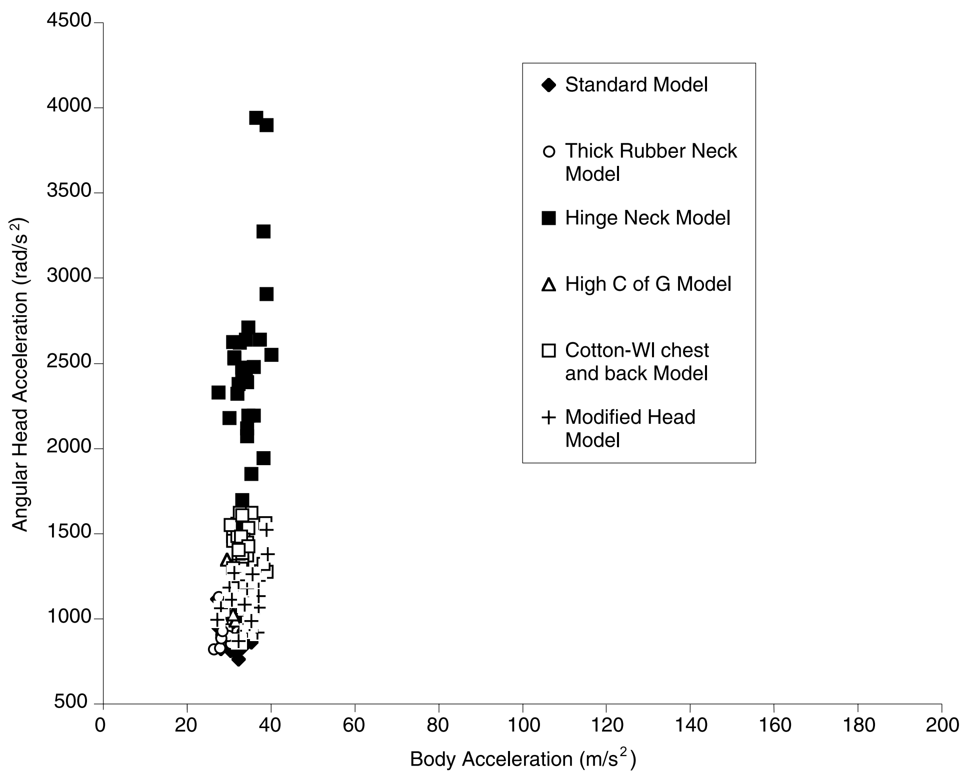

Figure 1. Adjustable model results showing effect of each parameter change on angular head acceleration (rad/s^2).

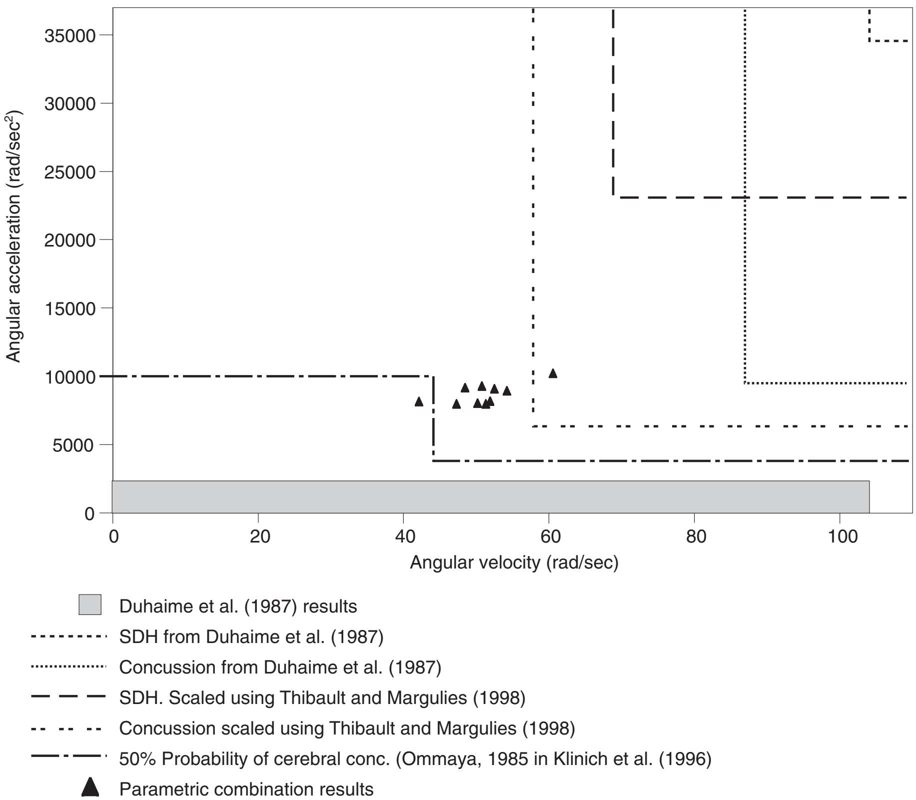

Figure 2. Shaking test results for parameter combination model showing comparison with Duhaime et al. (1987) results and tolerance limits for concussion and Subdural Haematoma (SDH) scaled for a one-month-old infant.

velocity (rad/s). The shaded region on the graph indicates the range of results from the original 1987 Duhaime et al. study. The internal head injury tolerance limits used by Duhaime for concussion and subdural haematoma are shown along with more recent tolerance limits suggested by Thibault and Margulies (1998). Also, another tolerance limit for 50% probability of onset of cerebral concussion (Ommaya, 1985) scaled using the method documented by Klinich et al. (1996), for a one-month-old's head dimensions is shown. All tolerance values were derived from animal studies.

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

RELEVANT BIOMECHANICAL RESEARCH FOR ASSESSING SHAKEN BABY SYNDROME SIMULATION RESULTS

Our preliminary studies show that shaking the model produced both chin-to-chest and back of head (occiput)-to-back impacts. If this were to occur in infants during shaking, that is, if it is anatomically possible, it will have profound implications to the argument that currently ensues as to whether pure shaking alone can produce fatal head injury, since it introduces a series of end-point impacts.

Other evidence of end-point impacts

Other studies have reported end-point impacts. Janssen et al. (1991) conducted physical crash tests using a TNO (the Netherlands Organisation for Applied Scientific Research)

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.

	Results for model head						Results for model body		
Test-Name	Peak tangential head AccN (m/s^2)	Peak angular head AccN (rad/s^2)	Peak tangential head velocity (m/s)	Peak angular head velocity (rad/s)	Time duration of peak head AccN curve (ms)	Head Injury Criterion (HIC)	Peak body AccN (m/s^2)	Peak body velocity (m/s)	Time duration of peak body AccN curve (ms)
Para-1	1385.47	8149.80	7.16	42.13	13	684	71.51	3.53	103
Para-2	1577.35	9278.53	8.63	50.78	11	1001	91.33	6.18	129
Para-3	1558.02	9164.85	8.24	48.47	11	952	96.14	5.89	121
Para-4	1736.86	10216.83	10.30	60.59	15	1421	73.97	4.02	144
Para-5	1515.65	8915.56	9.22	54.24	17	1039	67.98	4.02	120
Para-6	1366.93	8040.74	8.53	50.20	17	823	54.74	4.61	161
Para-7	1389.10	8171.15	8.83	51.94	18	866	79.66	5.69	174
Para-8	1353.00	7958.80	8.04	47.32	18	735	66.81	5.49	136
Para-9	1542.92	9075.98	8.93	52.51	17	1008	71.22	5.98	143
Para-10	1353.58	7962.26	8.73	51.36	18	846	64.06	5.79	172
Mean	1488.16	8693.45	8.66	50.95	16	938	73.74	5.12	140

P 3/4 (nine-month-old) child dummy. High-speed film analysis of the tests showed that the dummy's chin impacted the upper torso. 'These impacts not only affected the accelerations of the dummy's head, but also influenced the induced neck forces'. In parallel with the experimental work, a 2D Mathematical Dynamics Model (MADYMO) of the P 3/4 (nine-month-old) dummy was used to obtain a better understanding of the response of the child physical dummy. 'Particular emphasis was placed on analysing the effect of chin-to-chest contacts...'.

It is noteworthy that end point impacts, chin-to-chest and occiput-to-back, will produce increased tensile neck forces. Recently published work by Geddes et al. (2001a) documents the disruption of the craniocervical junction from both pure shaking and shaken-impact scenarios. 'Our study shows that infants of two to three months typically present with a history of apnoea or other breathing abnormalities, show axonal damage at the craniocervical junction, and tend also to have a skull fracture [not in the 'shaken-only' cases], a thin film subdural haemorrhage, but lack extracranial injury'. All reported cases were fatal as the study was conducted on post-mortem subjects. Geddes et al. (2001b) suggest that 'it may not be necessary to shake an infant very violently to produce stretch injury to the neuraxis. It is true that the more vigorous the shaking, the greater the stretch that would take place at the extremities of movement, and the worse the damage produced'.

Duhaime (1996) suggested the 'impact' of the back of the head (occiput)-to-back as a possible injury mechanism in the shaken baby syndrome. Commenting 'that much of the injury, subdural bleeding, occurs because, when the back of head strikes, the open lambdoid suture in infants indents into the brain causing tissue strains at the posterior deep bridging veins (the vein of Gallen, internal cerebral veins and straight sinus). Much of the bleeding seen, especially in the posterior interhemispheric fissure, has nothing to do with angular acceleration/deceleration, but is due to failure of the deep draining veins in the centre of the brain caused by an occipital impact. If that is the case, everything that was done with respect to injury thresholds may be wrong. But it's still impact'.

Problems associated with scaling tolerance limits

It is not clear from Duhaime et al. (1987) exactly which studies were utilised to scale from primates to infants to obtain tolerance values for pure angular head acceleration. However, Gennarelli et al. (1982) and Thibault and Gennarelli (1985) were referenced in the study and both studies used apparatus that ensured a purely angular acceleration was imparted to the subject. Additionally, in a personal correspondence with S.S. Margulies (1999) it was suggested that Gennarelli and Thibault (1982) was used as the source data for the subdural haematoma tolerance limits in the Duhaime et al. (1987) study. Gennarelli and Thibault (1982) commented of the apparatus used in the experimental set-up that 'the system provides nonimpact-distributed inertial loading conditions so that the effects of acceleration are studied in isolation'. Gennarelli et al. (1982) utilised apparatus that 'provided a nonimpact, distributed acceleration load to move the head in a controlled pathway'.

In a series of primate experiments they utilised a controlled mechanism (head in helmet linkage system) for inducing rotational (angular) acceleration of the head. The tests purposefully prevented the possibility of contact between the chin-to-chest or occiput-to-back. Even if the head were allowed free motion (as in the sled tests, Ommaya et al., 1967; Ommaya and Hirsch, 1971), structural and anatomical differences between adult primates and human infants may significantly reduce confidence in a direct comparison; for example, the relatively large neck muscles of adult primates may passively reduce head rotation and thus may prevent or greatly reduce any contact of the chin or back of the head (occiput) to the body. Therefore, an impact of the chin-to-chest or occiput-to-back type mechanism may not be intrinsic in animal tests.

If an impact or series of impacts of this type were to occur during pure shaking, impact tolerance limits should be used, based on impact tests, rather than the currently applied angular acceleration (shaking) data. This point can be illustrated by comparing threshold values for these two mechanisms. For a pure 'impulsive/indirect' angular acceleration, scaled for a three-year-old's brain mass, values of 8140 rad/s^2 for 10 ms are applied as an injury threshold (scaled from Ommaya et al., 1967; Stürtz, 1980). However, for an impact-induced angular acceleration, values of 2008 rad/s^2 for 10 ms are applied (scaled from Ommaya and Hirsch, 1971; Stürtz, 1980). The different tolerance values reflect the fact that the brain is less tolerant to impact induced angular acceleration than it is to pure indirect angular acceleration.

Stürtz (1980) commented, of the direct values (induced by impact), scaled for three and six-year-olds, that 'because a child's skull is less rigid, the direct application of the force means a generally higher endangering of the child. Therefore, the derived values for children could still be considerably reduced'. With a one-month-old child this comment is even more relevant.

Duhaime et al. (1987) assessed their shaking and shaken impact results using a scale with tolerance limits for cerebral concussion, subdural haematoma, and diffuse axonal injury scaled from subhuman primates, for the brain mass of a one-month old infant. Duhaime et al. (1987) suggested 'a tolerance scale... has been developed for the subhuman primate by Thibault and Gennarelli (1985)'. The scaling relationship shown in Thibault and Gennarelli (1985) is referenced as 'the one proposed by Holbourn (1943)'.

Holbourn's (1943) method was also documented and discussed by Ommaya et al. in 1967 and in many further publications by Ommaya and other authors. Ommaya et al. (1967) commented of the method that it could be used for 'extending the results of experiments on concussion-producing head rotations on lower primate subjects to predict the rotations required to produce concussion in man'.

The scaling data was subsequently applied to the paediatric population by Stürtz (1980).

Recent research has raised questions about the applicability of the scaling laws when scaling from human adults to infants. Prange et al. (1999) investigated a number of assumptions made during Holbourn's original scaling (1943, 1956), documented by Ommaya et al. (1967). Discussing both their findings and those of Thibault and Margulies (1998), Prange et al. (1999) suggested that these 'data demonstrate that the assumption of identical material properties in the model and prototype used by Ommaya et al. (1967) does not hold true for scaling between paediatric and adult inertial head injury'. Additionally, they concluded that 'the geometry of the adult brain was also found to be significantly different from the infant brain. This shows that the assumption of similar geometry between infants and adults also fails'. The authors suggested that their research 'provides a foundation for the study of the unique etiology and pathophysiology of paediatric brain injury'.

Ommaya made a further assumption, that 'the skull is very stiff, such that deformations of the skull do not contribute heavily to the strains of the enclosed brain' (Ommaya et al., 1967). Problems may arise however, if this assumption is applied to the paediatric population since a more compliant paediatric skull may produce strains in the enclosed brain (Thibault and Margulies, 1998; Margulies and Thibault, 2000).

Another concern is that the tolerance limits used to assess engineering models of shaking are scaled from data collected during biomechanical research within the automotive industry, which are based on a single whiplash event or a single impact event. This may be quite unlike a shaking episode during child abuse, where there may be a greater number of shakes and impacts, the cumulative effects of which are not assessed within the tolerance limits scaled from animal surrogates and adult cadavers. The authors suggest that the possible breakdown in material properties of cranial bridging veins and nerve axons with repeated tensile and/or shear and/or compressive strains may increase the potential for injury, compared to that of a single insult.

It is also imperative to acknowledge that current biomechanical analyses are viewed only from a mechanical perspective and fail to consider any subsequent pathophysiological consequence which may result from a primary injury.

Therefore, when tolerance limits are scaled to assess simulations involving infant models, they do not assess the risk of fatality considering all possible mechanisms of injury for the paediatric population, only the effects a single insult of pure angular head acceleration, that is, one possible mode of injury.

Also, if experimentation were to show a significant impact at the chin-to-chest and occiput-to-back, the authors suggest that future tests should consider both the effects of pure angular acceleration and impact (linear) acceleration using a head impact model, for example the Head Injury Criterion (HIC). If the impact tolerance limit were surpassed, even though the pure angular tolerance limits were not, the implication may be that fatal head injury may occur in pure shaking from injury mechanisms (chin-to-chest, occiput-to-back impacts) not identified by Duhaime's study in 1987. The effects of multiple impacts/shakes cannot be properly assessed until further research is conducted in this area. However, the authors suggest the potential damage of multiple impacts/shakes of the same acceleration magnitude would at least be equal to, and possibly even greater than, a single insult.

Head Injury Criterion (HIC)

Prasad and Mertz (1985), compared a collection of skull fracture and brain injury data with their corresponding HIC values and suggested that at an HIC value of 1000 there is a 16% risk of life-threatening brain injury in an adult population.

Stürtz (1980) simulated ten pedestrian accidents involving children (age not specified) using an anthropometric dummy. The simulations resulted in an HIC value of 840 being suggested as the critical load value of a child's head, rather than the 1000 value applied to adults. A 'critical load value' was defined as 'the load on the body under which an initial considerable damage of the organism takes place – destruction of a cell; irreversible injury – for instance when bone fractures occur or primary organs rupture'.

Mode of shaking

In the Duhaime et al. (1987) study, models were 'held by the thorax facing the volunteer and were shaken in the anteroposterior (A-P) direction, since this is the motion most commonly described in the shaken baby syndrome'. It must be noted that any minor deviation in this shaking pattern would result in the head moving in directions other than in the A-P plane. Since the measurement apparatus used in the original study was capable of measuring only in the A-P plane, measurement and assessment of even minor perturbations would require measurement apparatus capable of measuring head acceleration in those other directions. For example, if evidence were provided which suggested that a child was held so that its side was facing the assailant and the head accelerated laterally, this alternative mechanism would have to be considered.

Gennarelli et al. (1982) found that the direction of head motion (with acceleration remaining constant within narrow limits) was an important factor in producing the brain injury, Diffuse Axonal Injury (DAI), in primates, commenting, 'of the sagittal [A-P] group, 85% had a good recovery, while of the lateral group 84% had persisting coma or severe disability'. Margulies and Thibault (1992) scaled DAI thresholds to lateral rotation for a 500 gm brain mass (i.e. a one-month-old) which indicate a lower tolerance to lateral rotational acceleration than the angular rotational acceleration also scaled to a 500 gm brain mass in the Duhaime et al. (1987) paper. Therefore, any deviation from the standard anteroposterior shake pattern could predispose an infant to a greater injury and, therefore, should be noted and included in any subsequent simulation and/or opinion.

For a thorough review and explanation of head impact injury models and tolerance values see Cory et al. (2001).

DISCUSSION

Effect of model parameter changes on ratio-results

- The ratio-results from the adjustable model tests showed that three factors, the metal hinge-neck, high centre of gravity and cotton wool chest and back padding parameters, significantly increase the ratio-results, compared to the standard model. These results emphasise a requirement that models for the investigation of the SBS simulate an infant as accurately as possible in terms of mass distribution (centre of gravity) and response (biofidelity of neck, chest and back).
- The three parameter changes, centre of gravity, neck type and chest and back padding, were adjusted on the same model and subjected to an increased shake effort and altered shake pattern. The shake tests produced results higher than those produced in the original Duhaime et al. (1987) study, also surpassing two scaled tolerance limits for concussion. On a cautionary note, evidence has arisen from literature that raises doubts about the validity of scaling tolerance limits from primates and adults to infants. It is possible that these limits are inaccurate to some unknown degree. As the results from the parameter combination model are closer to the limits (for subdural haematoma) than the Duhaime model results, this factor becomes more significant in the issue of whether pure shaking alone is capable of causing fatal head injuries to infants.

In the Discussion and Conclusion the 'ratio-results' are defined as dimensionless values, indicating a comparable level of angular head acceleration for all parameter changes.

Impact of chin-to-chest and back of head (occiput)-to-back

- The contact between the chin-to-chest and occiput-to-back of the model during shaking tests indicates that there may be another injury mechanism in 'pure shaking', not previously investigated. Evidence (Ommaya and Hirsh, 1971; Stürtz, 1980) suggests that if an impact occurs, before rotational (angular) acceleration, the head injury tolerance may be reduced, i.e. less additional rotational acceleration may be required, in addition to the impact, to cause a fatal brain injury.

- In light of the fact that there is an impact between the chin-to-chest and occiput-to-back during the shake tests conducted in this study, the angular acceleration tolerance limits may not be applicable in all shaking scenarios. The controlled mechanism of inducing the rotational (angular) acceleration of the head, in the previously mentioned primate tests, was designed to purposefully prevent the possibility of contact between the chin-to-chest or occiput-to-back. Since impact was clearly identified in the experimental series, for the parameter combination model, the results were assessed using an impact based tolerance limit (Head Injury Criterion (HIC)). Forty per cent of the results were over the HIC threshold of 1000 and 80% were over the HIC critical loading value of 840 suggested by Stürtz (1980) for children.

The term shaken baby syndrome (SBS) defines the assailant's actions on the model/infant, but it does not adequately describe all the possible mechanisms of injury. The term SBS has evolved to incorporate shaking and head impact(s) with an external object or surface. This study has suggested a series of impacts occur between the head and body during pure shaking, a back of head (occiput)-to-back impact, then a whiplash of the head, followed by a chin-to-chest impact. It may be possible to induce a purely angular acceleration of the head using different shaking mechanisms. However, it is suggested that in a frenzied attack (as simulated in the parameter combination tests), with many shake cycles, it is highly likely that endpoint impacts would occur. We are primarily concerned, in this study, with the question of causing *fatal head injury* during 'pure shaking' and therefore the main emphasis is on the worst case scenario.

It is suggested that when the 'pure shaking' of an infant is described in a biomechanical context a better term, which adequately describes the most likely mechanism, would be impact-whiplash-impact (IWI). The authors would like to note that no suggestion is being made that this be added to the already abundant descriptions of the SBS entrenched in the medical literature. It is suggested that the term described above is more useful in the description of the biomechanical mechanisms present in violent 'pure shaking' scenarios.

Problems with scaling

- There is some evidence from literature that the assumptions made originally to scale from animals to adult humans do not necessarily apply to scaling from adults to infants. It is suggested that if these assumptions are not valid, when the scaling law is applied between human adults and human infants (Prange et al., 1999), they must also be invalid when scaling directly from primates to human infants.
- Thibault and Margulies (1998) have put forward a new method for scaling, based on the results of recent research on paediatric (surrogate) brains. These changes lower the tolerance limits for infants.
- The results were compared with tolerance limits from the automotive industry where the effects of a single event (whiplash and/or impact) were measured, rather than the multiple insults associated with the SBS. The cumulative effect of many shakes and/or impacts is unknown.

CONCLUSION

The 1987 Duhaime et al. study documented a biomechanical simulation of the shaken baby syndrome. During the study angular head accelerations surpassing fatal head injury tolerance limits could not be produced in infant models from shaking alone. The simulation was conducted over 16 years ago and although it was a valuable first step to a better understanding of the biomechanics of the shaken baby syndrome, because it has been widely applied in the courts and literature, the model was in need of re-assessment.

To test the validity of the model a preliminary study was undertaken in which a replica was constructed and tested. The vague description of the design and construction of the Duhaime model in the original paper allowed for the possibility that variations might be introduced during its reproduction. It was postulated therefore, that differences in certain parameters might affect angular head accelerations.

The authors conducted a parametric study with an adjustable replica of the Duhaime model to answer the following question:

> *'What factors significantly affect the angular head acceleration of a biomechanical model constructed to simulate the shaken baby syndrome?'*

Experimental results suggested that three parameters significantly increased the level of angular head acceleration: metal hinge-neck, high centre of gravity and cotton wool chest and back material.

These findings emphasise the requirement that future models of the shaken baby syndrome accurately simulate an infant, in terms of the biomechanics of mass distribution (centre of gravity) and response (biofidelity of neck, chin-to-chest and occiput-to-back contact points).

The angular head acceleration and velocity results from the parameter combination model, with increased shake effort and altered shake pattern, surpassed the Duhaime et al. (1987) results and spanned two scaled tolerance limits for concussion. As this adjustable replica 'Duhaime' model produces different acceleration values from the original study, it is evident that changing certain parameters affects angular head acceleration. However, it cannot be claimed that either model (i.e. the original Duhaime et al. (1987) model or the parameter combination replica Duhaime model) is biofidelic. It is currently unknown whether an improved level of biofidelity in some parameters would increase or decrease the angular head accelerations produced during pure shaking. However, it can be suggested that if these parameters do affect the results they must be designed to be as biofidelic as possible for reliable conclusions to be drawn.

The model produced in 1987 was very simplistic and was not designed to resemble a human infant in terms of mechanical response (biofidelity). In the last 16 years much research has been conducted in the area of modelling children using mechanical crash test dummies and computer models. It is now known that to run meaningful simulations, models must be based on appropriate data, to design and assess (calibrate) the head, neck, and thorax for biofidelity. The Duhaime et al. (1987) study has been widely quoted (197 times, Science Citation Index 2003) in other papers on the subject and is often quoted in SBS litigation. However, although it has been criticised in the literature it has not previously been properly reproduced and systematically assessed. The authors suggest that the current study has provided evidence to suggest that changes in the biomechanical properties of the model influence the results for angular head accelerations. Neglecting these factors produces a model with an unknown resemblance to an infant, therefore, any simulation results obtained with such a model will be meaningless and conclusions drawn unreliable.

The conclusions drawn from the current study emphasise the need for the design and construction of a biofidelic infant model to simulate shaking before results can be reliably quoted in the literature and/or applied in a court of law. However, there are many other problems associated with the Duhaime et al. (1987) biomechanical study and future studies on the biomechanics of the shaken baby syndrome:

- Evidence suggests that the tolerance limits used to assess the shaking simulation results in Duhaime et al. (1987) may not be reliable. The degree to which they are inaccurate is unknown.
- The results of all shake tests conducted during this study identified clear impacts at the chin-to-chest and occiput-to-back sections of the shake cycle. Therefore, the (possibly inaccurate) tolerance limits utilised by Duhaime et al. (1987) may not be applicable in the assessment of shaking simulations due to the impacts identified in this study.

- Although the current model was not biofidelic, the impact tolerance limits suggested for children were surpassed in 80% of the parameter combination model shake test results which indicates that endpoint impacts, if identified in future tests, should be assessed with impact-based tolerance limits.
- Even though 80% of the results surpass the impact tolerance limit the cumulative effect of multiple impacts (from many shake cycles) cannot be assessed as the tolerance limits are based on single impacts (as in car crash scenarios). Therefore, although some shaking/impact results might be below fatal head injury tolerance limits, the effect of repeated consecutive sub-lethal loading is unknown.

It is suggested that further research into the design, construction and assessment of a model for SBS research is required to develop a biofidelic infant model, in light of the research conducted and child data published since the Duhaime et al. (1987) study.

The application of data from animal surrogate experiments, adult cadaver experiments and scaling calculations has greatly assisted in overcoming the problems associated with the paucity of child data, when developing child-safe environments. However, extreme caution should be exercised when applying the data in a medico-legal context. In addition, since the Duhaime model has an unknown level of biofidelity, presentation of the study in evidence in any criminal prosecution runs the risk of its prejudicial effect outweighing its probative value and may result in any arbiter of fact wrongly interpreting the evidence.

Therefore, at this present stage, the authors conclude that it cannot be categorically stated, from the Duhaime et al. (1987) study, that 'pure shaking' cannot cause fatal head injuries in an infant. There must, therefore, be sufficient doubt in the reliability of the Duhaime et al. (1987) biomechanical study to warrant the exclusion of such testimony in cases of suspected shaken baby syndrome.

REFERENCES

Alexander R., Sato Y., Smith W. and Bennett T. (1990) Incidence of impact trauma with cranial injuries ascribed to shaking. *AJDC* **144**, 724–6.

Beusenberg M.C., Happee R., Twisk D. and Janssen E.G. (1993) Status of injury biomechanics for the development of child dummies. *SAE* Paper No 933104.

Bruce D.A. and Zimmerman R.A. (1989) Shaken impact syndrome. *Pediatr. Ann.* **18**, 482–94.

Bundez D.L., Farber M.G., Mirchandani H.G., Park H. and Rorke L.B. (1994). Ocular and optic nerve hemorrhages in abused infants with intracranial injuries. In: Gilliland M.G. and Folberg R. (1996). Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Caffey J. (1972) On the theory and practice of shaking infants: Its potential residual effects of permanent brain damage and mental retardation. *AJDC* **124**, 161–9.

Cory C.Z. (2001) A biomechanical assessment of the Duhaime shaken baby syndrome model. Doctor of Philosophy Thesis, University of Wales, Cardiff, U.K.

Cory C.Z., Jones M.D., James D.S., Leadbeatter S. and Nokes L.D.M. (2001) The potential and limitations of utilising head impact injury models to assess the likelihood of significant head injury in infants after a fall. *Forensic Sci. Int.* **123**, 89–106.

Duhaime A.C., Gennarelli T.A., Thibault L.E., Bruce D.A., Margulies S.S. and Wiser R. (1987) The shaken baby syndrome: A clinical, pathological, and biomechanical study. *J. Neurosurg.* **66**, 409–15.

Duhaime A.C. (1996) Research on the pathophysiology of the shaking-impact syndrome. *The Pediatric Trauma and Forensic Newsletter* **4**, 73–7.

Duhaime A.C. (1998). Personal Correspondence.

Duhaime A.C., Christian C.W., Rorke L.B. and Zimmerman R.A. (1998) Current Concepts: Nonaccidental head injury in infants – The 'shaken-baby syndrome'. *N. Engl. J. Med.* **338**, 1822–9.

Elner S.G., Elner V.M., Arnall M. and Albert D.M. (1990). Ocular and associated systematic findings in suspected child abuse. In: Gilliland M.G. and Folberg R. (1996) Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Geddes J.F., Hackshaw A.K., Vowles G.H., Nickols C.D. and Whitwell H.L. (2001a) Neuropathology of inflicted head injury in children: I. Patterns of brain damage, *Brain* **124**, 1290–8.

Geddes J.F., Vowles G.H., Hackshaw A.K., Nickols C.D., Scott I.S. and Whitwell H.L. (2001b) Neuropathology of inflicted head injury in children: II. Microscopic brain injury in infants, *Brain* **124**, 1299–306.

Gennarelli T.A. and Thibault L.E. (1982) Biomechanics of acute subdural hematoma. *J. Trauma* **22**, 680–6.

Gennarelli T.A., Thibault L.E., Adams J.H., Graham D.I., Thompson C.J. and Marcincin R.P. (1982) Diffuse axonal injury and traumatic coma in the primate. *Ann. Neurol.* **12**, 564–74.

Gilliland M.G. and Folberg R. (1996) Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Guthkelch A.N. (1971) Infantile subdural haematoma and its relationship to whiplash injuries. *BMJ* **2**, 430–1.

Hadley M.N., Sonntag V.K.H., Rekate H.L. and Murphy A. (1989) The infant whiplash-shake injury syndrome: a clinical and pathological study. *Neurosurgery* **24**, 536–40.

Holbourn A.H.S. (1943) Mechanics of head injuries. *Lancet* 438–41.

Holbourn A.H.S. (1956) Private Communication, 13th Oct. to Dr. Sabina Strich. In: Ommaya A.K., Yarnell P., Hirsch A. and Harris E. (1967). Scaling experimental data on cerebral concussion in subhuman primates to concussive thresholds for man. *Proc. 11th Stapp Car Crash Conf. Anaheim, CA.*

Ingraham F.D. and Matson D.D. (1954). Neurosurgery of Infancy and Childhood. In: Caffey J. (1972). On the theory and practice of shaking infants: Its potential residual effects of permanent brain damage and mental retardation. *AJDC* **124**, 161–9.

Irwin A.L. and Mertz H.J. (1972) Biomechanical bases for the CRABI and Hybrid III Child Dummies. *Child Occupant Protection, 2nd Symposium Proceedings P-316, Orlando, Florida, SAE*, November 12, Paper No. 973317.

Jacobi G. (1986). Damage patterns in severe child abuse with and without fatal sequelae. In: Gilliland M.G. and Folberg R. (1996). Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Janssen E.G., Nieboer R., Verschut R. and Huijskens C.G. (1991) Cervical spine loads induced in restrained child dummies. *35th Stapp Car Crash Conf., P-251, SAE,* Paper No, 912919.

Klinich K.D., Saul R., Auguste G., Backaitis S. and Kleinberger M. (1996) Techniques for developing child dummy protection reference values. *Report by the child injury protection team, NHTSA*, October 1996.

Margulies S.S. (1999) Personal Correspondence.

Margulies S.S. and Thibault L.E. (1992) A proposed tolerance criterion for diffuse axonal injury in man. *J. Biomechanics* **25**, 917–23.

Margulies S.S. and Thibault K.L. (2000) Infant skull and suture properties: Measurements and implications for mechanisms of pediatric brain injury. *J. Biomechanical Engineering* **122**, 364–71.

Massicotte S.J., Folberg R., Torczynski E., Gilliland M.G.F. and Luckenback M.W. (1991) Vitreoretinal traction and perimacular retinal folds in the eyes of deliberately traumatized children. In: Gilliland M.G. and Folberg R. (1996). Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Munger C.E., Peiffer R.L., Bouldin T.W., Kylstra J.A. and Thompson R.L. (1993) Ocular and associated neuropathologic observations in suspected whiplash shaken infant syndrome. In: Gilliland M.G. and Folberg R. (1996). Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Norris B. and Wilson R. (1995) CHILDATA, The handbook of child measurements and capabilities – data for design safety, *DTI Consumer Safety Unit*, June 1995, ISBN 0-9522 571-1-4.

Ommaya A.K., Yarnell P., Hirsch A. and Harris E. (1967) Scaling experimental data on cerebral concussion in subhuman primates to concussive thresholds for man. *Proc. 11th Stapp Car Crash Conf. Anaheim, CA.*

Ommaya A.K. and Hirsch A.E. (1971) Tolerance of cerebral concussion from head impact and whiplash in primates. *J. Biomechanics* **4**, 13–21.

Ommaya A.K. (1985) Biomechanics of head injury: experimental aspects. In: Nahum A.M. and Melvin J. (eds.) *The Biomechanics of Trauma, N.J. Prentice-Hall*. pp.245–79.

Pheasant S.T. (1986) Bodyspace: anthropometric ergonomics and design. In: Norris B. and Wilson R. (1995). CHILDATA, The Handbook of Child Measurements and Capabilities – Data for Design Safety, *DTI Consumer Safety Unit*, June 1995, ISBN 0-9522 571-1-4.

Prange M.T. and Margulies S.S. (1999) Pediatric rotational brain injury: The relative influence of brain size and material properties. *Proc. of 43rd Stapp. Car Crash Conf, SAE*. pp 333–41.

Prasad P. and Mertz H.J. (1985) The position of the United States delegation to the ISO Working Group 6 on the Use of HIC in the Automotive Environment. *SAE*. Paper No. 851246.

Rao N., Smith R.E., Chou J.H., Xu X.H. and Kornblum, R.N. (1988) Autopsy findings in the eyes of fourteen fatally abused children. In: Gilliland M.G. and Folberg R. (1996). Shaken babies – some have no impact injuries. *J. For. Science* **41**, 114–6.

Snyder R.G., Schneider L.W., Owings C.L., Reynolds H.M., Golomb D.H. and Schork M.A. (1977) Anthropometry of infants, children, and youths to age 18 for product safety design, UM-HSRI-77-17 Final Report Contract CPSC-C-75-0068 to *Consumer Product Safety Commission*, May 31. http://www.itl.nist.gov/div894/ovrt/projects/anthrokids/anthrokids.html

Steenbekkers L.P.A. (1993) Child development, design implications and accident prevention, No.1 in Physical Ergonomics Series. In: Norris B. and Wilson R. (1995). CHILDATA, The Handbook of Child Measurements and Capabilities - Data for Design Safety, *DTI Consumer Safety Unit*, June 1995, ISBN 0-9522 571-1-4.

Stürtz G. (1980) Biomechanical data of children. *24th Stapp Car Crash Conf. Proc., SAE*. Paper No. 801313.

Thibault K.L. and Margulies S.S. (1998) Age-dependant material properties of the porcine cerebrum: effect on pediatric inertial head injury criteria. *J. Biomechanics* **31**, 1119–26.

Thibault L.E. and Gennarelli T.A. (1985) Biomechanics of diffuse brain injuries. In: *Proc. of the Fourth Experimental Safety Vehicle Conference*. New York, American Association of Automotive Engineers.