PROBABILITY OF PEDIATRIC SKULL FRACTURE AT VARIOUS CONTACT VELOCITIES

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ABSTRACT

The main objective of this research was to quantify the probability of simple linear skull fracture in infants under the age of 6 months when the head contacts a rigid surface at a range of velocities. Probability was quantified by conducting drop tests using the Aprica 2.5 dummy. Other objectives of this study were; to develop a methodology that can be used to relate dummy and infant cadaver head impact response; to quantify the effects of dummy body orientation on head acceleration biofidelity; to identify pediatric head structures that need to be modelled in a child dummy; and finally, to identify a child dummy head design which is likely to provide biofidelic head acceleration over a range of impact velocities. Aprica 2.5, a 2.5 kg instrumented infant dummy was dropped from heights of 0.376 m to 3 m onto rigid plates. Isolated head and whole body dummy drop tests were conducted. Dummy head impact response, when appropriately scaled, is very similar to infant cadaver head response for rigid plate contact. Contact velocities in this study ranged from 2.3 m/s to 6.3 m/s. The probability of linear skull fracture ranges from ≤ 5% for contact velocities to ≤ 3 m/s and < 50% for contact velocities ≤ 5.5 m/s. Results confirm the validity of 5% fracture tolerance limit for CRABI – 6M dummy obtained previously through mass and material scaling.

Work presented in this paper indicates that a one-piece moulded dummy head, such as the one in Aprica 2.5 may preserve biofidelity over a range of impact velocities. This work provides insights for FE modelers and dummy designers about the importance of various skull model parameters. Such a dummy design with human like moments of inertia when used in conjunction with a human like head – neck connection may also provide reasonable estimates of infant head angular accelerations.

INTRODUCTION

In this paper, we will estimate probability of non-displaced skull fracture in infants (0-6 months age) when the head contacts a rigid surface. Traditionally, injury has been related causally to impact through cadaver tests. However, societal and ethical concerns have restricted pediatric cadaver testing. Limited isolated infant cadaver head testing has been conducted by Prange, (2003) and Loyd, (2011). They dropped isolated heads of infant cadavers onto rigid plates and measured head acceleration and force on the plate. Loyd (2011) analyzed the data extensively and proposed methods to correlate child dummy head acceleration with infant cadaver head response.

Weber, et al. (1984, 1985) conducted full body child cadaver drop tests. The age of test subjects varied from neonate to 9 months. All subjects were dropped from a height of 82 cm with the test subject horizontal at the time of release. They dropped cadavers onto rigid and padded surfaces. All test subjects dropped onto rigid surfaces sustained non-displaced skull fractures some of which crossed suture lines. The cadavers used by Weber were not instrumented.

Snyder et al (1963, 1977) attempted to estimate the relationship between injury severity, fall height, and type of fall surface. They documented falls from heights up to 11 m. They used a combination of detailed medical and scene investigations and computer modeling to relate fall heights and injury for free falls on to rigid surfaces. Samuel, et al. (2015) conducted a single center cohort study at a pediatric emergency department (ED) of a Level I trauma center over a period of 2 years. Study participants were children from 0 to 2 years of age who were admitted to the ED due to minor head injury. Ibrahim (2009) conducted a retrospective study of infants ($0 \leq$ age ≤ 12 months) and toddlers (12 months $\leq$ age ≤ 48 months) hospitalized for head trauma sustained as a result of falls.

Li, et al. (2015) developed a finite element model to analyze falls reported by Weber et al. (1984, 1985). They developed one isolated head model to represent each Weber drop test. Each head model captured the head circumference of the test subject it was modeling and was also developed to be age appropriate. They related probability of skull fracture to engineering variables such as peak head linear acceleration, strain, etc.

Van Ee, et al. (2009) reproduced Weber's tests using a CRABI (Child Restraint / Air Bag Interaction} 6-month-old dummy. They developed a probability curve relating dummy head acceleration and skull fracture.

Other researchers have conducted drop tests with child dummies to evaluate the risk of injury from falls (Bertocci, et al. (2003, 2004)). In addition to assessing biomechanics associated with short-distance falls in children, they also investigated the effect of impact surface type on injury risk.

Coats (2008) developed an anthropometric infant dummy and dropped it from heights of 1 ft, 2 ft, and 3 ft onto concrete, carpet pad and mattress so that the dummy contacted the floor in a supine position. She reported that peak head force increased with increase in stiffness of the floor and that drop height had a significant effect on impact force.

Some researchers have developed finite element models of pediatric heads to study the effect of falls (Coats (2003), Ibrahim (2009), Klinich (2002), Roth (2008). There is some debate about material properties of model elements and the importance of sutures.

Two recent reports (Ruddick, et al. (2009) and Monson, et al. (2008)) discuss infant in-hospital falls. Ruddick reported that 14 neonates fell from heights ranging between 0.3 m and 1.09 m. Five of the neonates fell onto rigid floors from heights ranging from 0.5m to 1.2m. Ruddick reported six infants who sustained linear skull fractures. However not all children in the study were scanned. Monson indicated that that one of the 14 infants who fell in nurseries sustained a skull fracture. Not all children in the study were scanned for skull fractures. Both studies reported no mortalities or adverse neurologic outcomes from the falls. While these studies do not provide a causal relationship between skull fractures and measured engineering variables, they provide useful real-life data which can be used to validate skull fracture probabilities developed in this study.

When a relatively small, light, soft object such as an infant head contacts a relatively rigid surface of much larger dimensions, it is possible to use the principles of Hertzian contact mechanics to evaluate the force imposed on the head and its acceleration on impact. In such head contacts, the pulse width (the time difference between the first and last contacts between the surface and head) is nearly constant regardless of the fall height. Rangarajan, et al (2013) used this principle of near invariance of pulse width (NIP) to analyze the relationship between fall height and injury in a 3-month-old child who was taken to a pediatric ED after a fall.

Infants sustain head injury from falls and in motor vehicle accidents and there is a need to evaluate probability of skull fracture from both these causes of injury. Prior efforts have related peak head linear acceleration, which is a dependent variable, with probability of skull fracture. These formulations require that tests be conducted with dummies or cadavers before probability of fracture is determined. However, it would be useful to be able to estimate probability of skull fracture without having to conduct experiments.

In many circumstances, velocity of contact is the easiest independent engineering variable to estimate in falls and motor vehicle accidents. In the case of falls, contact velocity can be reasonably accurately related to fall height.

Traditionally, infant dummy head response biofidelity is evaluated by conducting dummy and infant cadaver head impacts under the same conditions. For example, if an isolated cadaver head is dropped from a certain height, then, isolated dummy head would be dropped from the same height onto the same surface. Head accelerations would then be analyzed to evaluate dummy head response biofidelity and to estimate scale factors if needed. However, dummy tests in this study were conducted much before cadaver head drop tests were conducted. The tests were run in two different institutions and were not coordinated. Therefore, we have used non-traditional analytic methods to compare dummy response with cadaver response. For example, isolated cadaver head was dropped from a height of 30 cm leading to a contact velocity of 2.2 m/s whereas isolated dummy head response was evaluated at a contact velocity of 2.7 m/s. In addition, we compared response of the dummy head in a whole body vertex impact at a contact velocity of 6.3 m/s with isolated cadaver head response at the same velocity onto the similar rigid plate. These non-traditional analytic raise a number of questions which are discussed in the DISCUSSION section.

In this study we conducted drop tests with an instrumented infant dummy and used the NIP principle to analyze test results and to evaluate the probability of an uncomplicated, linear non-displaced skull fractures from impact velocities up to 6.3 m/s

METHODS AND MATERIALS

Our aim was to evaluate the probability of moderate head injury such as non-displaced linear skull fracture caused by head impact against a rigid surface at various contact velocities for infants (age ≤ 6 months). The Aprica 2.5 kg instrumented dummy was used to as an infant surrogate in our tests. The work was divided into two phases.

- Conduct dummy isolated head drop tests, analyze data by scaling the peak head acceleration and pulse width and establish a procedure for comparing dummy and infant head impact responses. Pulse width used for scaling was obtained from literature describing infant cadaver isolated head drop tests. We established the viability of the scaling procedure and the range of biofidelity of the dummy.
- Dummy was dropped on its vertex from 1 m to 3 m heights. Dummy head accelerations for various contact velocities were compared with previously published relationships between infant head acceleration and skull fracture probability to establish probability of skull fracture at a range of contact velocities.

Methods and materials – features of the Aprica 2.5 dummy

Figure 1: Aprica 2.5 Dummy

Aprica 2.5 instrumented dummy (Figure 1) was used to conduct drop tests. Design of the dummy was described in Rangarajan (2002). Salient details from the paper are reproduced here for completeness.

The dummy structure consists of head, neck, thorax (flexible thoracic spine and shoulder structure), pelvis, and lower and upper extremity segments. The segmentation scheme is similar to that seen in adult dummies and was chosen as it allowed the infant dummy to be instrumented with a suite of sensors similar that seen in adult dummies.

Dummy anthropometric measurements are shown in Table 1 as is a list of sensors integrated in the dummy. Preliminary anthropometric data were obtained from anthropometric specifications for a premature infant in CMVSS (Canadian Motor Vehicle Safety Standards) 213.5 test procedures (Transport Canada, 2000). Mass scaling procedures proposed by Mertz (1989) and Melvin (1995) were used to adjust the dimensions of the infant data in CMVSS 213.5. Data were also obtained by pediatricians in Japan who measured several body segments of Japanese infants weighing approximately 2.5 kg. Dimensions obtained by pediatricians were compared with the segment dimensions published by the Japanese Ministry of Transport for a 10% infant of mass 2.5 kg.

Table 1: Aprica 2.5 anthropometry and instrumentation

Parameter	Units	Design Goal	Measured
Mass	gm	2600	2600
Height	mm	450	450

Parameter	Units	Design Goal	Measured
Head circumference	Mm	349	340
Neck length	M	54	53
Shoulder circumference	Mm	305	340
Chest circumference	Mm	297	298
Waist circumference	Mm		318
Hip circumference	Mm	286	285
Leg length (crotch to heel)	Mm	150	133
Segment Mass			
Head weight	Gm	800	772
Neck weight	gm	126	62
Torso (shoulder, thorax, pelvis)	gm	1273	1244
Upper arm weight	gm	29	39
Lower arm weight	gm	22	32
Upper leg weight	gm	82	79
Lower leg (w/ foot) weight	gm	48	73
Integrated sensors Type	**Location**		
3-axis accelerometer	Head CG		
3-axis accelerometer	Neck top		
3-axis accelerometer	Neck bottom		
3-axis accelerometer	Mid-torso, T4		
3-axis accelerometer	Pelvis		

Methods and material – dummy head design

Material properties of H3 50th percentile male dummy head were measured and scaled using procedures outlined by Melvin (1995). The head was molded from 30 Shore A Durometer 2-part Urethane. Urethane's density is very close to water and it is a lightly viscous material. The head was attached to the neck by the occipital condyle (OC) bolt. Accelerometers located at the CG of the head were attached to a Delrin mount that was used to locate the OC bolt. This type of head construction allowed the head to deform uniformly. Also, since the head was molded from a single material, its material properties were the same all over the head.

Methods and material - dummy neck design

The neck was fabricated from a Urethane tube with fixtures at the top and bottom to house tri-axial accelerometers. The neck was connected to the head through the OC bolt. The neck was designed to fit the scaled Mertz corridor using procedures outlined by Melvin (1995). The biomechanical response of the Aprica 2.5 fits into the scaled Mertz corridor. The neck is stiff in bending compared to static bending requirements described by Coats (2008) but its bending properties are similar to those used by Ibrahim (2009) in the 18-month-old child dummy.

Methods and material - dummy thorax and abdomen design

The torso flesh was molded in one piece from 30 Shore A Durometer Urethane. Foam was placed inside the thorax to model deformable thoracic and abdominal structures. Thoracic spine was fabricated from a Urethane tube and was quite flexible. The spine was divided into two parts along its length and tubes of appropriate lengths were connected together through an aluminum plate. A tri-axial accelerometer package was attached to the aluminum plate.

Tests indicated that the thorax was a little stiffer in compression than the scaled Kroell (1969) response requirements. Calibration tests of the abdomen indicated that peak impact force of 190 N was within the expected range of 187-230 N obtained by scaling rod impact tests from Cavannaugh (1987).

Methods and material – details of dummy tests

Before using a dummy to simulate human drop tests, it is important to understand the limits of the biofidelic response of the dummy. In this case, the limit of biofidelic response will be defined as the range of contact velocities (proportional to drop heights) over which the dummy head acceleration is very similar to that of human surrogate of the same age and anthropometry when exposed to similar test conditions.

Two series of tests were conducted (Table 2). Limits of biofidelic response were obtained by analyzing series 1 tests. Series 2 tests were analyzed to evaluate skull fracture probability at various contact velocities when the head contacted a rigid surface.

In all tests, dummy responses were recorded at 10 KHz sampling rate, together with high-speed video at 500 f/s. Dummy acceleration data were filtered using standard SAE J 211 CFC 1000 filters.

Table 2: Type of free fall tests

Series No.	Test type	# of Tests	Drop Hts, m	Details of tests
A	1. Forehead impact	4	0.376	Isolated dummy head dropped onto a rigid steel plate.
	2. Vertex impact	2	2	Whole dummy dropped on its vertex onto steel plate.
B	Vertex impact	10	1, 1.5, 2.5, 3, 3.5	Whole dummy dropped on its vertex onto steel plate from 6 different heights. 2 tests per height.

Methods and material – Isolated head drop tests

In these tests, the head of the dummy was isolated from the dummy. The head was suspended 376 mm above a rigid steel plate. The setup used the standard H3 head calibration drop test fixture. The head was released from that height and dropped under gravitational acceleration to strike a rigid steel plate. Three accelerometers placed at the center of gravity of the head, and pointing in the front-back, lateral, and inferior-superior direction recorded the accelerations in the respective directions. By design, the head rebound after the first impact was minimal as can be seen in Figure 3.

Methods and material – Vertex drop tests

Figure 2: Vertex Drop Test Setup

Occipital condyle bolt on the dummy (connecting top of the neck to the head) was set to just support the head. All joints were set so that they just supported the body segment below them under gravitational load. For example, the shoulder joint was set so that the arm was just retained in the position it was placed in. These procedures are commonly used to prepare crash test dummies for tests.

In these tests, the dummy foot was loosely tied to a lanyard and suspended head down from the test fixture so that the vertex was at the test height from the rigid platform. The lanyard was released so that vertex of the dummy contacted the rigid platform as shown in Fig. 2. Kinematics of the dummy were repeatable and the vertex contacted the rigid plate first (before the body) in all tests.

Methods and material – calculation of contact velocity from drop heights

Contact velocity was calculated in all tests from drop heights neglecting frictional effects. Contact velocity (Vel) in m/s is related to drop height by:

$$\text{Vel} = \sqrt{(2 * 9.81 * \text{drop height (m)})} \quad \text{(Equation 1)}$$

RESULTS

Results of tests conducted to establish limits of biofidelity of head impact response will be discussed first, followed by results of vertex drop tests.

Results – Series A tests to establish limits of head impact response biofidelity

Limits of dummy head biofidelity were established by comparing dummy head response with infant cadaver response (Loyd, 2011) under similar loading conditions. Initially, dummy head response in isolated head drop tests were compared with appropriate infant cadaver test response. Following this, whole dummy vertex drop test in which the head contacted a rigid plate at 6.26 m/s (2 m drop height) was compared with infant isolated head drop test from a height of 2 m (Loyd, 2011).

Resultant head acceleration versus time plot in four isolated head drop tests is shown in Fig. 3 which indicates that dummy response was repeatable. Figure 4 shows the mean peak head acceleration and pulse width. Pulse width is the difference in time from the time the head contacts the plate to the time the head bounces off the plate. The mean head response is 115 G with a pulse width of approximately 7.9 ms.

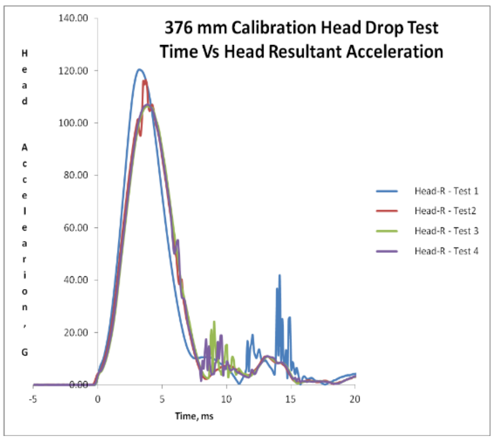

Figure 3: Dummy head acceleration in four tests contact velocity 2.7 m/s.

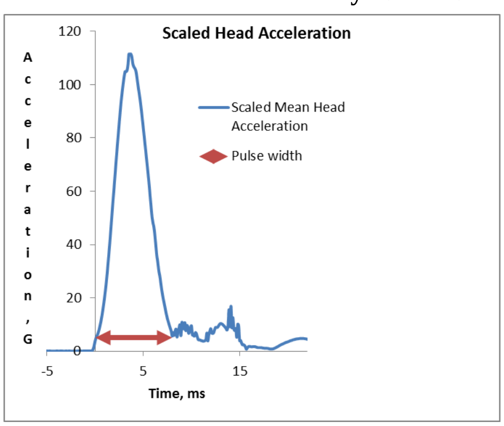

Figure 4: Dummy mean head acceleration.

Head impact response was scaled using the principle that in impacts between head and a defined surface, pulse width does not change appreciably with contact velocity or head mass (Rangarajan, et al. 2013). The comparison process consists of morphing the head acceleration curve by attenuating acceleration amplitude and stretching time by a given scale factor. This type of scaling causes the area under the scaled curve to be the same as the area under the original curve. In other words, initial velocity and initial energy of the scaled pulse are the same as that of the unscaled cadaver acceleration pulse. A similar process has been proposed by Loyd to adjust CRABI-6M dummy response to better fit human response. (Loyd, 2011)

The scale factor equals dummy response pulse width / infant cadaver response pulse width from infant cadaver tests of the same age group. Pulse width from infant cadaver tests was obtained from Loyd (2011) and has a value of 17 ms. So, the pulse width scale is 7.9 / 17 = 0.46.

Scaled head response pulse is shown in Fig. 5 together with the unscaled pulse and response of one of the cadaver heads in Loyd's tests. Scaled dummy peak acceleration is around 54 G which is close to the 58 G average reported by Loyd (2011). It is to be noted that contact velocity was 2.4 m/s in Loyd's tests and 2.7 m/s in our test.

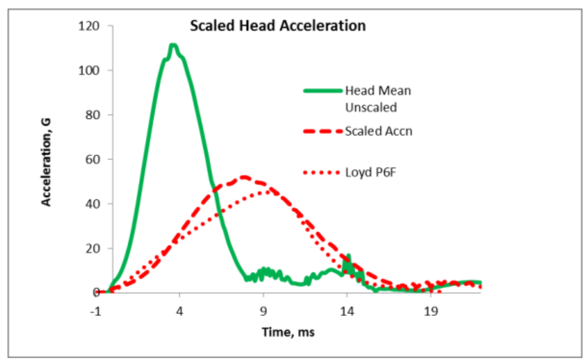

Figure 5: Comparison of scaled and unscaled dummy, and representative infant cadaver P6F from Loyd (2011)

Rate of loading in the dummy is very similar to that of neonates. This fact is significant because the rate of loading has been recognized as an important variable that can predict injury severity (Snyder, 1977). The shape of the unloading portion of the dummy response is similar to that of the neonate which might indicate that the type of material used to mold the head (2-part Urethane) is a reasonable choice. Scale factor and the results of scaling are shown in Table 3.

Table 3: Comparison of Aprica 2.5 adjusted mean head resultant acceleration with Loyd (2011) data.

Data Source	Mean Peak, G	Scale factor
Loyd (2011) average neonatal forehead drop, contact velocity 2.4 m/s	58	
Aprica scaled mean forehead drop, contact velocity 2.7 m/s	53	0.46

The 2.7 m/s isolated head drop tests provided the lower limit of biofidelity of dummy head impact response. To obtain the upper level of biofidelity we chose to compare dummy response from 6.3 m/s drop test using an infant cadaver (P12M – 5-month-old, head mass 0.96kg from Loyd (2011)) with 6.3 m/s dummy head contact velocity test using the NIP scaling technique described above. Figure 6, shows head acceleration traces obtained by Loyd (2011) in a 6.3 m/s infant head drop and the scaled head acceleration curve for 6.3 m/s test of the Aprica 2.5 dummy whose head mass is 0.8 kg. Once again, the amplitudes and shape of curves are reasonably well correlated and loading rate is very similar for the two subjects. Loyd (2011) reported a peak acceleration of 159 G and the scaled dummy peak acceleration is 155 G, roughly 3% lower than infant cadaver test acceleration.

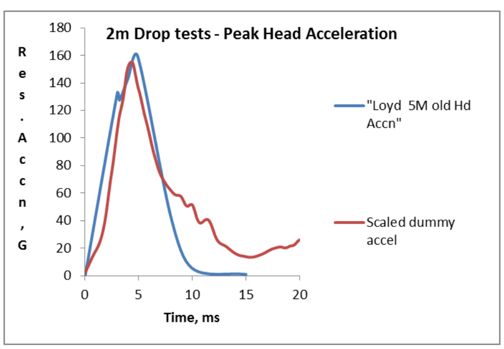

Figure 6: Comparison of scaled Aprica and infant cadaver head accelerations in 6.3 m/s contact velocity tests

Results from the 2.7 m/s isolated head drop test and 6.3 m/s vertex drop test indicate that the dummy will yield reasonably biofidelic results for this range of contact velocities if dummy response is mapped appropriately.

Results – Series B vertex drop tests

Dummy response in 1, 1.5 and 2 m vertex drop tests was then analyzed. Scaled accelerations and drop heights from vertex drop tests number 7 through 9 are listed in Table 4.

Table 4 Peak resultant head accelerations in vertex impact tests

Test No	Drop height. mm	Contact Vel, m/s	Peak head acceleration, G	Scaled Pk. Hd. acceleration, G
7	1000	4.4	180	85
8	1500	5.4	236	111
9	2000	6.3	330	155

Results – probability of skull fracture at various contact velocities

In the last few sections, we estimated head CG acceleration resulting from contact with a rigid surface at various velocities. We then used these data in conjunction with the analysis of Li, et al. (2015) to estimate the probability of skull fracture due to these contacts.

Li, et al. (2015) developed finite element models of the infant head to simulate Weber's whole body infant cadaver drop tests. Their models varied by age and geometry (mainly head circumference). Li et al. (2015) used this model to estimate variables commonly associated with skull fracture such as peak head acceleration, von Mises stress, and head Injury criterion (HIC). Li, et al. (2015) related probability of linear skull fracture for ages ranging from 0 to 6 months, which is of interest to the present work. These probability values are listed in Table 5.

Table 5: Estimates of age-based infant skull fracture probability from Li, et al (2015)

Approximate Infant Peak Head Acceleration, G (0 m to 6m age)	Approximate Calculated Probability of Skull Fracture, %
84 -92	5
106 –114	25

Approximate Infant Peak Head Acceleration, G (0 m to 6m age)	Approximate Calculated Probability of Skull Fracture, %
119 -127	50
132 – 140	75
144 – 153	90

We compared the dummy peak head linear accelerations for various contact velocities (Table 4) and assigned probabilities of fracture. For example, the isolated head impact test with a contact velocity of 2.7 m/s resulted in a peak linear head acceleration of 52 G. From Table 5, there is a < 5% probability that this would cause linear skull fracture. So, this acceleration level was assigned a linear skull fracture probability of <5%. Dummy head response from our drop tests are tabulated below with estimates of fracture probability.

Table 6: Summary of Aprica 2.5 drop test results

Contact Velocity, m/s	Type of test	Scaled dummy response, G	Skull Fracture Probability from Li, et al.
2.7	Series A– Isolated Head Drop	53	≤ 5%
4.4	Series B – Vertex Drop Test	85	≤ 25%
5,4	Series B – Vertex Drop Test	111	≤ 50%
6.3	Series A – Vertex Drop Test	155	90%

DISCUSSION

As discussed in the Introduction, this study uses non-traditional analytic methods to establish dummy head response biofidelity which leads to a number of questions about the methodology. Some of these are:

- Is it appropriate to compare dummy whole body tests with isolated head drop tests conducted by Loyd (2011)?
- Is it appropriate to use average pulse width for forehead impacts from Loyd (2011) to scale vertex drop tests using Aprica 2.5 dummy?
- Is the scaling procedure capable of estimating head acceleration at contact velocities not tested in our study? If the procedure is appropriate it should be able to estimate infant head accelerations at contact velocities that were not tested using the Aprica 2.5 dummy. In other words, if the scaling procedure is used to estimate infant head acceleration at 1.7 m/s (15cm drop test) and 2.4 m/s (30 cm drop test), how close will these estimates be to Loyd’s (2011) experimentally observed infant head accelerations for the same drop heights.

These questions will be discussed in the next three sections.

Appropriateness of comparing whole body and isolated head drop tests

Our results indicate that the soft infant head deforms first upon contact and that the torso mass undergoes negligible deceleration during the time head peak acceleration attains a maximum value. This is illustrated by Fig. 7 which shows a plot of the Z (downward) acceleration of the head, and torso in 2 tests with contact velocities of 4.4 m/s and 8.3 m/s. This plot covers the range of contact velocities investigated in this study. The same sequence of deceleration is seen when head resultant decelerations are compared in other vertex drop tests.

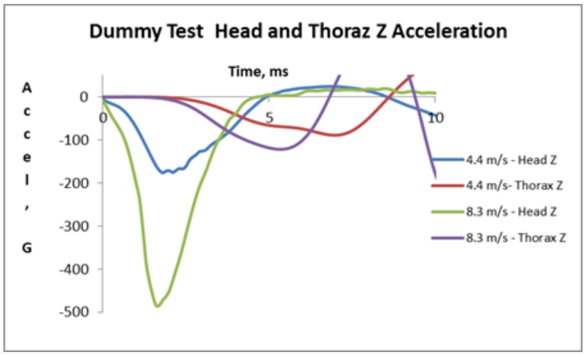

Figure 7: Sequence of vertical deceleration of head and thorax segments in 4.4 m/s and 8.3 m/s contact velocity tests

Figure 7 shows that at both contact velocities, the head starts decelerating first and comes to a stop before rebounding. During the period the head is decelerating, the torso does not undergo significant deceleration. In essence, this indicates that the head is the only active mass involved in the impact. Thus, these tests are similar to isolated head drop tests and it is appropriate to compare whole body dummy vertex drop test results with infant isolated head drop tests.

The torso accelerometer is placed approximately at the CG of the torso. This sequence of deceleration can be explained by the softness of the head (Shore A 30 durometer Urethane), and the flexibility of the relatively long, thin tube used to model the thoracic spine. Therefore, the torso accelerometer which is place approximately at the CG of the long, slender thoracic spine starts decelerating after the head comes to a stop.

Appropriateness of using average forehead impact pulse width to scale vertex impact data

Table 7 lists pulse width data abstracted from Loyd (2011). To populate the table, six infants under the age of six months (P3, P5, P6, P7, P8, and P13 in Loyd (2011)) were chosen. Their respective head masses were 0.42, 0.61, 0.65, 0.42, 0.68 and 0.45 kg. Pulse width for each infant in each impact direction was averaged. Finally, the average of these averages was calculated across all six infants. The average of these averages was calculated to be 17.26 ms as indicated. This is very close to the 17 ms which was used as the initial estimate scaling factor based on cadaver test pulse head impact pulse width.

Table 7: Average pulse width for each impact location from Loyd (2011)

Impact location	**Mean +/- SD pulse width of 6 infants, ms**
Vertex	18.68 +/- 4.7
Occiput	17.49 +/- 4.9
Forehead	16.54 +/- 5.6
Right parietal	16.8 +/- 4.4
Left Parietal	16.8 +/- 3.4
Average of averages	**17.26 +/- 0.87**

Appropriateness of scaling procedure

Dummy scaled resultant head acceleration for 2.7 m/s, 4.4 m/s, 5.4 m/s and 6.3 m/s tests were 53 G, 85 G, 111 G and 155 G respectively. These data were analyzed by fitting polynomial, exponential and linear curves to these data. The regression curves were extrapolated to estimate scaled dummy head resultant acceleration at 1.7 m/s and 2.4 m/s which were the contact velocities in Loyd (2011). Result of this analysis is presented in Table 8 which lists estimated values of scaled head acceleration with experimental infant cadaver head acceleration reported by Loyd (2011). It is seen that estimates from both polynomial and exponential regressions are quite close to experimental head accelerations. This shows that it is possible to extrapolate dummy impact test results to estimate infant head impact accelerations.

Table 8: Comparison of estimated scaled dummy head acceleration and experimental infant cadaver test data

Contact Vel, m/s	**Loyd Average head accel, G**	**Polynomial, R^2 - 0.99**	**Exponential, R^2 = 0.99**	**Linear, R^2 = 0.98**
1.7	39	48	48	36
2.4	58	52	51	45

Data shown in Table 8 indicates that the scaling factor used and the concept of scaling using appropriate cadaver pulse width to scale dummy head response data yields reasonable results.

How do estimated probabilities compare with literature reported value?

Table 6 indicates that a scaled dummy head acceleration of 53 G is associated with less than a 5% probability of skull fracture. This value is very similar to that derived by Melvin, (1995) and Mertz et al., (1984, 1984b, 1989) as the tolerance value for the CRABI – 6M dummy. Melvin (1995) scaled adult dummy tolerance values using both mass and material properties to arrive at a tolerance value for the dummy. Our results in Table 6 support his approach.

Summary of results

We used an instrumented Aprica 2.5 infant dummy weighing 2.5 kg to conduct whole body drop tests onto a rigid platform from heights ranging from 1 m to 3 m. We also conducted isolated head drop tests onto a rigid platform. Dummy head accelerations were scaled using a procedure outlined by Rangarajan (2013). The scaling procedure provided a method to compare the dummy response with isolated human infant head responses under similar test conditions.

From the results of this study, we conclude that:

- Pulse width scaling is a reasonable procedure to evaluate limits of biofideloty of dummy responses.
- Aprica 2.5 dummy yielded biofidelic head impact responses for impacts against rigid surfaces in the contact velocity range of 2.7 m/s to 6.3 m/s when scaled appropriately.
- The average of infant cadaver pulse width for vertex, occiput, forehead, and left and right parietal contacts reported by Loyd (2011) can be used to scale dummy response.
- Analysis in this paper suggests it is possible to derive a simple, back-of-the-envelope calculation for probability of infant skull fracture for various contact velocities (fall heights) against rigid surfaces.
- Within the range of contact velocities analysed in this paper, it is appropriate to conduct dummy whole body tests with Aprica 2.5 and other similarly designed dummies with biofidelic heads and flexible thoracic spines to mimic isolated human cadaver head tests under similar test conditions.
- The Aprica 2.5 head was moulded from a single material – Urethane. Such a design is likely to provide biofidelic head impact responses for various head impact locations against rigid surfaces.
- This simple design of the Aprica 2.5 dummy head can provide important information about material properties of the skull and brain for finite element model developers.
- Urethane density is similar to that of the brain biologic material. Therefore, a one-piece moulded design, such as the anthropometric Aprica dummy head, is likely to have human-like moments of inertia in all directions. This in turn is likely to result in accurate estimates of angular acceleration of the head in various directions. Tests conducted with such dummies will provide useful information about head angular accelerations which can be related to real life intracranial injuries as angular accelerations and angular velocities have been corelated with brain injury.
- A one-piece moulded dummy head may preserve biofidelity over a range of impact velocities. Design of such a dummy head with instrumentation has been illustrated. This work provides insights for FE modelers about the importance of various skull model parameters.

LIMITATIONS

- Biofidelity of dummy head acceleration was established at two discrete contact velocities. It is necessary to establish that the dummy response is biofidelic at all velocities between these limits by conducting appropriate cadaver tests.
- Dummy drop test data indicate that the head completes deformation and deceleration before the torso starts decelerating. It is necessary to investigate if this is true in children

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