Probability model relating contact velocity and pediatric head injury severity

Nagarajan, Rangarajan
Tariq, Shams
GESACInc.com
Tsuguhiro, Fukuda
Consultant
Japan

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ABSTRACT

The objective of this study was to use epidemiologic, and infant cadaver drop test data to develop a probabilistic model relating probability of non-displace skull fracture to contact velocity for infants aged up to 6-months. A secondary objective was to verify the accuracy of mass and material scaling methods used in the past to develop head injury tolerance criteria for CRABI-6M dummy. Infant fall data reported in the literature were combined with infant cadaver drop test data to develop a data set of 80 head impacts. Contact velocity for each impact in the data set was estimated from drop height; and head acceleration was estimated using pulse width from infant cadaver drop tests. Estimated peak head acceleration was related to probability of skull fracture. Estimated probability was compared with pediatric skull fracture probabilities reported in literature. The curve relating contact velocity with linear skull fracture has the form

$$P = e^{(-6.5199 + (1.5658V_c))} / (1+ e^{(-6.5199 + (1.5658V_c))})$$

Where V_c is the contact velocity, which in this study, ranges from 1.7 m/s to 4.9 m/s. Probabilities estimated in this study agree with previously reported values thus validating the calculation procedures used in this study.

INTRODUCTION

Falls and motor vehicle accidents are an important cause of pediatric Emergency Department (ED) visits (Marin, et al. 2014). However, there is a lack of information about tolerance levels for various types of head injuries in infants. Traditionally, cadaver tests have been used to relate head impact injury caused by a fall. 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 infant cadaver heads onto a rigid plate. Weber, et al. (1984, 1985) conducted full body child cadaver drop tests. They dropped uninstrumented cadavers onto rigid, and padded surfaces. All children dropped onto rigid surfaces sustained simple linear skull fractures.

Two recent reports (Ruddick, et al. (2009) and Monson, et al. (2008)) discuss infant in-hospital falls. These authors reported on infant fall onto rigid hospital floors from heights ranging from 0.5m to 1.2m. Ruddick reported a number of linear skull fractures whereas Monson indicated that that only 1 of the 14 infants sustained a skull fracture. Presence or absence of skull fracture was not confirmed in all cases in both studies.

Snyder et al (1963, 1977) documented falls from heights up to 11m in an attempt to estimate the relationship between injury severity, fall height, and type of contact surface. They used a combination of detailed medical and scene investigation, and computer modeling to relate fall heights and injury for free falls on to surfaces of varying stiffness.

Outcomes of falls in the pediatric population has been studied either through retrospective studies of hospital admissions (Ibrahim, 2009) or using finite element models (Coats, 2003, Ibrahim, 2009, Klinich, 2002, Roth, 2008) or through dummy drop tests (Bertocchi, et al. (2003 and 2004), Coats, 2003).

Li, et al. (2015) developed a finite element model to analyze falls reported by Weber (1984, 1985). They related peak head linear acceleration to probability of skull fracture. Van Ee, et al. (2009) conducted drop tests using CRABI (Child Restraint – Air Bag Interaction) 6-month old dummy to reproduce Weber’s drop tests. They developed a curve relating peak head acceleration to probability of pediatric skull fracture. The fall height in both these studies was set at 0.81m to match the Weber’s (1964, 1985) study.

Rangarajan, et al. (2013) noted that pulse width of adult cadaver head impacts is very weakly related to drop height for a given contact surface. The drop height in these tests varied from 0.6m to 2.1m. This increase of 250% in drop height caused approximately 7% decrease in the pulse width. They used this observation to calculate peak head acceleration of one child brought to the ED with a simple linear skull fracture.

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. Prior efforts have related peak head linear acceleration, which is a dependent variable, with probability of skull fracture. Rangarajan (2017) related scaled peak head acceleration of a biofidelic infant dummy head to fracture probability at three discrete fall heights (proportional to contact velocities). Li, et al (2015) and van Ee (2009) related skull fracture probability to peak head acceleration at a fixed fall height (proportional to contact velocity). To our knowledge, a continuous curve relating probability to skull fracture for various contact velocities (fall heights) is not available at present.

Additionally, relations between probability of fracture and head acceleration require that tests be conducted with dummies or cadavers before such a relation can be developed. Head acceleration is a dependent variable in an impact in the sense that impact causes head acceleration. However, in many cases of falls and motor vehicle accidents, it is not too difficult a task to estimate contact velocity which is an independent variable. A probability relationship between skull fracture and an independent variable will be a very useful tool allowing researchers to develop initial estimates of skull fracture probability without having to conduct tests or to develop and exercise complicated models.

In this paper, we estimate peak linear head acceleration using the procedure described by Rangarajan, et al. (2013) for infant fall cases available in literature. Literature used in this study listed the fall height, and described contact surface and consequent injuries in each fall. We then related the peak head accelerations to moderate head injuries (non-displaced skull fractures) through a probability curve. Both fracture and non-fracture cases available in literature for impacts against a rigid surface were used in our analysis. The relationship between probability of skull fracture and peak head acceleration was then converted to relationships between probability and contact velocity and / or fall height using procedures developed by Rangarajan (2013).

METHODS AND MATERIALS

Our objective is to develop a probabilistic relationship between contact velocity (related to fall height) and simple linear skull fracture in infants (age ≤ 6 months) for falls onto rigid surface. The process of development of the probability relationship was divided into the following steps:

1. Develop a formula relating head contact velocity and fall height.
2. Obtain Pulse Width for infant falls onto rigid surfaces from Loyd (2011). Pulse Width is defined as the difference in time between the 1st contact of the head with the rigid surface and the beginning of the first rebound. During this period, the head deceleration goes from zero to maximum and goes back to zero. A typical head impact pulse and pulse width are shown in Fig. 1.
3. Develop a list of infant fall cases described in literature where the falls surface was rigid and fall heights and outcome injuries were known.
4. Estimate peak head accelerations for all study cases and relate measured and estimated accelerations to probability of simple linear skull fracture. Procedure used by Rangarajan, et al. (2013, 2017) was used to estimate peak head acceleration.

Details of the four steps are provided below.

Calculation of head contact velocity

Neglecting air friction, contact velocity "V_c in m/s" of the head at the end of a fall of "h" meters under gravitation forces is given by

$v = \sqrt{2gh}$ (Equation 1)

Where "g" is the gravitational constant and has a value of 9.81 m/s^2

Obtain pulse width for infant cadaver isolated head drop tests

Loyd (2011) and Prange (2004) conducted a number of infant cadaver head drop tests from 0.15m and 0.3m heights. The average Pulse Width (PW) in Loyd's tests was 17 ms for forehead drops onto rigid surfaces for the age group of interest (0 ≥ age ≥ 6 months). Analysis by Rangarajan, et al. [2017] established that:

- It is appropriate to use the forehead drop test pulse width for infant falls where Vertex, Occiput, and left and right parietes make first contact with a rigid surface for the velocities of interest in this study.
- It is appropriate to use pulse widths from isolated head drop tests for full body falls in dummies similar to Aprica 2.5 infant dummy.

Pulse width average of 17.26 ms calculated in Rangarajan [2017] will be used in this study.

Estimate peak head acceleration

We used the Impulse-Momentum theorem which is obtained by rearranging terms in Newton's second law of motion (Force = Mass * Acceleration). If force F applied to a body of mass M causes a change in velocity ΔV during time T, then, Newton's second law can be stated as follows:

$F * T = \mathrm{M} * \Delta\mathrm{V}$ (Equation 2)

Or, Impulse = Change in Momentum

Substituting (Force = Mass * Acceleration), we obtain

$Acceleration = \Delta\mathrm{V}/\mathrm{Time}$ (Equation 3)

When a head contacts a rigid surface, it starts decelerating till its velocity is zero. Deceleration reaches a maximum value when the head velocity is zero. Most damage to the head occurs during this deceleration or loading phase when head velocity goes its initial velocity to zero.

To simplify calculation, we can assume that the deceleration – time curve is triangular in shape. This assumption is supported by Fig. 1 which shows a reconstruction of measured head acceleration profile from one of Loyd's (2011) infant cadaver head drop tests. It is seen that during the loading phase, acceleration increases linearly from zero (at the time the head contacts the surface) to a maximum value approximately midway through the pulse width. For this shape of deceleration pulse, the average acceleration is ½ of the peak value. So, we can assume that a constant acceleration (average acceleration) is applied from the time the head contacts the surface to the end of loading phase midway through the pulse width. Equation (3) now reduces to:

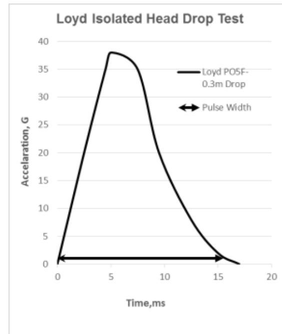

Figure 1: Cadaver head drop test from Loyd

$PHA = 2 * \mathrm{Vc}/(0.5 * PW)$ (Equation 4)

$PHA = 106.2 * \sqrt{\mathrm{h}}$ (Equation 5)

Where:

PHA = Peak Head Acceleration in G, m/s^2.

PW = Pulse width in milliseconds, as shown in Fig. 1. From Loyd's data, we determined that PW averages to 17ms for infants (age ≤ 6-months) for 0.15m and 0.3m fall heights.

h = height in meters.

Obtain infant fall data from literature

We used data from Loyd (2011), Monson (2009), Rangarajan (2013), Ruddick (2008), and Weber (1984, 1985) to develop a probability curve relating PHA and Probability of moderate skull fracture. Data used to develop the probability curve are summarized in Tables 1 and 2.

Loyd (2011) listed drop height and measured head peak acceleration for each test. Weber (1984, 1985) dropped uninstrumented child cadavers onto rigid and non-rigid surfaces from a fixed height (81 cm) and we calculated peak head acceleration using Equation 5. Monson, et al. (2008) and Ruddick, et al. (2009) listed drop heights and we calculated peak head accelerations using Equation 5. Rangarajan, et al (2013) provided drop height and peak head acceleration (Please note that there was a calculation error in Rangarajan (2013), the corrected peak head acceleration is presented in this table).

Monson (2008), Rangarajan (2013), Ruddick (2009), and Weber (1984, 1985) provided details of injuries. Ruddick (2009) reported that a number of subjects in her study were not scanned and they are not included in this analysis. Similarly, patients who were not scanned in the Monson (2008) study are not included in this analysis. In addition, one patient in the study sustained a depressed fracture. No further information was available, so this infant was not included in the analysis.

Loyd (2011) conducted 5 drop tests from 2 drop heights (0.15m and 0.3m) on 7 subjects within the age group we are considering (0 - 6months). In these 70 (7*5*2) tests, he reported one parietal fracture of neonate (subject P12M). Peak head acceleration data from all 70 tests were used in the analysis.

Table 1: Data used to develop probability of AIS2 skull fracture relationship with peak acceleration

Data source	Injury Description from data source	# of tests
L	Parietal fracture. Subject P12M, 0.15m drop	1
L	No fracture. Tests with subjects P03M, P05F, P06M, P07M, P08M, P12M and P13F.	69
Ra	Simple linear parietal fracture	1
Ru	No clinical signs, right parietal fracture	1
Ru	No clinical signs, left parietal fracture	1
Ru	No clinical signs, right parietal fracture, ultrasound normal.	1
Ru	Swelling in left parietal areas, left parietal fracture, ultrasound normal	1
Ru	Traumatic encephalopathy, right fronto-parietal fracture, cerebral contusion	1
Ru	No imaging, no clinical signs	1
Ru	No imaging, no clinical signs	1
Ru	No imaging, no clinical signs	1
Ru	No imaging, no clinical signs	1
Ru	No imaging, no clinical signs	1
Ru	No imaging, no clinical signs, bruise over temporal bone	1
We	Simple linear fracture	1
M	No fracture, no scan	1
M	No fracture, no scan	1
M	No fracture, no scan	1
M	No fracture, no scan	1
M	No fracture, CT Scan	1
M	No fracture, skull radiograph	1
M	No fracture, no scan	1
M	No fracture, skull radiograph	1
M	No fracture, no scan	1
M	No fracture, skull radiograph	1
M	No fracture, skull radiograph	1
M	No fracture, no scan	1
M	No fracture, no scan	1
M	No fracture, no scan	1

Table 2: Continuation of Table 1

Data source	# of tests	Fall Height, m	Peak head acceleration Estimated (E) or measured (M), G
L	1	0.15	41, M
L	69	0.15 and 0.3	26 to 112, M
Ra	1	1.3	120, E
Ru	1	0.5	75, E
Ru	1	1.0	106, E
Ru	1	0.5	75, E
Ru	1	0.5	75, E
Ru	1	1.2	116, E
Ru	1	0.8	95, E
Ru	1	1.0	106, E
Ru	1	0.5	75, E
Ru	1	0.5	75
Ru	1	0.5	75
Ru	1	0.5	75
W	1	0.81	96, E
M	1	0.8 – 1.1	111 E
M	1	0.8 - 1.1	111 E
M	1	0.8 – 1.1	111 E
M	1	0.8 – 1.1	111 E
M	1	0.66	87E
M	1	0.30	59 E
M	1	0.91	102
M	1	1.09	111
M	1	1.01	107
M	1	0.81	96
M	1	1.09	111
M	1	0.91	102
M	1	1.09	111
M	1	0.91	102

Keys to Column 1 (Tables 2&3):

1. L = Loyd,

2. Ra = Rangarajan,
3. Ru = Ruddick,
4. M = Monson,
5. W= Weber

In Tables 1 and 2, we have included all cases reported by Ruddick (2009) and Monson (2008) including infants who were not scanned. However, we included in our analysis only cases where presence or absence of skull fracture was confirmed by scans. We decided not to include Monson and Ruddick cases without scans for the following reasons:

- All other cases reported included scans to confirm skull fracture.
- Distribution of unscanned cases indicates that roughly the same numbers, about 8, would be included in the analysis above and below the 50% probability level. Thus, the distribution would not be very different from the one where no unscanned data were included.

RESULTS

We used data from Loyd (2011), Monson (2008), Rangarajan (2013), Ruddick (2009), and Weber (1984, 1985) to develop a probability curve for moderate skull fracture. The probability curve developed is shown in Fig. 2.

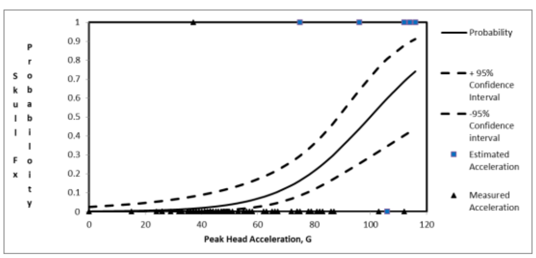

Figure 2: Probability of AIS 2 head skeletal injury related to peak head acceleration

The equation of the probability curve is

$$P = e^{(-6.5199 + (0.06528*PHA))} / (1 + e^{(-6.5199 + (0.06528*PHA))})$$
(Equation 6)

Where PHA = Peak Head Resultant Acceleration from Equation (5).

Substituting for PHA in Equation (6) results in Equation (7) that relates contact velocity and probability of moderate skull fracture.

$$P = e^{(-6.5199+(6.93*\sqrt{(\text{fall height})})} / (1+ e^{(-6.5199+(6.93*\sqrt{(\text{fall height})})})$$
(Equation 7)

Estimate of injury probability obtained from this curve is similar to previously reported estimates as seen in Table 4.

For easier comprehension, Table 3 below reproduces data from Fig.2. The second column shows the probability for each head acceleration listed in the first column. Columns 3 and 4 show the estimated contact velocity and estimated fall height respectively for the level of head acceleration listed in Column 1 by applying Equations 4 and 5.

Table 3: Height of fall, Contact Velocity, Peak Acceleration and Skull fracture probability.

Estimated Peak Head Accel, m / s^2	**Skull Fracture Probability**	**Estimated contact velocity, m/s**	**Est. Fall Height, m**
41	0.02	1.72	0.15
58	0.06	2.42	0.3
75	0.17	3.13	0.5
101	0.51	4.20	0.9
106	0.60	4.42	1
116	0.74	4.85	1.2
121	0.80	5.05	1.3
130	0.88	5.42	1.5
184	1.00	7.47	3

Comparison of current study results with literature

In the past, Mertz et al. [1986], Melvin [1995], Klinich [2003], van Ee et al. [2002], Coats [2002], and Li et al. [2015] have used various processes to relate head peak acceleration to probability of skull fracture. Results of our study are compared to the results of these studies to evaluate the appropriateness of our methodology. This comparison is presented in Tables 4 and 5.

Table 4: Comparison of study estimates with literature values

Source	Risk level of skull Fx	Process	Drop Height Repor-ted	Con-tact Sur-face
Mz,	5%	Scaling mass and material properties	N/A	N/A
Me	5%	Scaling mass and material properties	N/A	N/A
K	50%	Finite element model. Crash test	N/A	N/A
V	50%	CRABI drop tests. Vary contact surface	0.8m	Stone tile, carpet
C	50%	Finite element model		
L	5%	Finite element model. Varied contact surfaces	0.8m	Stone tile, carpet
L	50%	Finite element model. Varied contact surfaces	0.8m	Stone tile, carpet
Cu	5	Algebraic formula – current study	0.15 m to 1.2m	Rigid steel plate
Cu	50%	Algebraic formula – current study	0.15 m to 1.2m	Rigid Steel plate

Table 5: Continuation of Table 4

Source	Acceleration	
	Newborn	6-month old
Mz,		156
Me	69	67
K		85
V		82
C	29-35	
L	84	93
L	119	127
Cu	55	
Cu	101	

Key to Column 1 (Tables 4&5):

1. Mz = Mertz, et al (1986)
2. Me = Melvin (1995)
3. K = Klinich, et al (2003)
4. V = van Ee et al (2002)
5. C = Coats (2007)
6. L = Li et al (2015)
7. Cu = Current study

The probability curve in Fig. 2 and data in Tables 4 and 5 indicate that our estimates are close those provided by other researchers. However, previous probability curves (Li, (2015), Van Ee (2009)) have been constructed with data from one fall height. Our probability curve is designed to handle all fall heights up to 1.2m and directly relates level of injury to fall height and contact velocity. To the best of our knowledge, this is the only effort, apart from that of Snyder (1977) to relate fall heights to injury severities. Subjects in Snyder's study generally fell more than 3m and the youngest subject was 13 month old.

CONCLUSIONS

1. A probability curve constructed using a mixture of estimated and measured peak head accelerations for falls less than or equal to 1.2 m is comparable with those constructed using complex finite element models and dummy drop tests. Thus, it seems feasible to use the proposed algebraic formula to estimate skull fracture probability for contact velocities less than 4.3 m/s.
2. Finite element models yield very detailed information about the fracture, and response of the brain that is not provided by the proposed algebraic model. However, finite element models of child head require a large amount of data to define the geometry of the head, and material properties of the brain, skull, scalp, and sutures. It has been hard to generate these data given restrictions in child cadaver and child cadaver tissue testing.
3. Instrumented dummies are generally expensive and testing with them requires expensive ancillary equipment such as data acquisition hardware and software. Dummies also require periodic calibration using specialized equipment. So, both dummy testing and finite element models require more effort than most busy medical centers can afford to invest. We are hopeful that the simple analytic procedure discussed in this paper will encourage researchers to collect data from a large number of fall cases that come to the ED. Since the proposed model requires only 2 pieces of information from patients – fall height and a detailed list of injuries on the first visit to ED, we are hopeful that more researchers will collect information about a lot of falls thus making the proposed model more robust. Such a robust model can be used by finite element modelers to refine their models and conduct in-depth investigations into the effect of falls.
4. Linear acceleration of the CG of the head which is the output of the proposed model is related arithmetically to angular acceleration and angular velocity of the head by the formula: **Head angular acceleration = Head linear acceleration / radius of rotation.** Center of rotation in infants is not known but it has been estimated to be around C2, or about 1/4 the length of the neck. This simple formula can be used to relate intracranial injuries to angular acceleration in a large number of cases thus forming the basis of an investigation into the effect of impact on MTBI and TBI.
5. This work verifies the appropriateness of material and geometric scaling techniques proposed by Melvin (1995)
6. The analytic method used in this study can be expanded to older children and used to design better pedestrian head impact protection for children.
7. These results provide guidance for the development of test devices to model child head impact and a reduction in the need for child cadaver tests.

Limitations

1. The proposed probability curve does not separate effects of falls onto various parts of the head. However, analysis presented in Rangarajan (2017) indicates All impacts used in this study are against rigid surfaces. There is a need to extend this work to other surfaces such as seat back cushions, carpets, soil, etc.
2. The procedure used assumes that deceleration of the rest of the body does not significantly affect peak linear deceleration of the infant head. While this is true of the Aprica 2.5 dummy (Rangarajan, et al (2017)), it may not be applicable to live infants.

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