Biomaterials and Biomechanics in Bioengineering, *Vol. 3, No. 1 (2015) 47-57*
DOI: http://dx.doi.org/10.12989/bme.2016.3.1.047

Sensitivity analysis of skull fracture

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(Received May 21, 2015, Revised November 24, 2015, Accepted December 4, 2015)

Abstract. Results from multiple high profile experiments on the parameters influencing the impacts that cause skull fractures to the frontal, temporal, and parietal bones were gathered and analyzed. The location of the impact as a binary function of frontal or lateral strike, the velocity, the striking area of the impactor, and the force needed to cause skull fracture in each experiment were subjected to statistical analysis using the JMP statistical software pack. A novel neural network model predicting skull fracture threshold was developed with a high statistical correlation (R^2=0.978) and presented in this text. Despite variation within individual studies, the equation herein proposes a 3 kN greater resistance to fracture for the frontal bone when compared to the temporoparietal bones. Additionally, impacts with low velocities (<4.1 m/s) were more prone to cause fracture in the lateral regions of the skull when compared to similar velocity frontal impacts. Conversely, higher velocity impacts (>4.1 m/s) showed a greater frontal sensitivity.

Keywords: biomedical engineering; bone biomechanics; mechanics coupled with human activity; modeling and simulation; medical mechanics

1. Introduction

Several modes of injury exist in the general term of “traumatic brain injury” (TBI). Symptoms of TBI and the changes that they make on a person’s way of thinking, emotion, and sensory symptoms can be difficult to spot, which has prompted the CDC to label TBI as a “silent epidemic”. In an official report to Congress in March 2015, the CDC further stressed the dangers of TBI and called for better initial detection methods Centers for Disease Control and Prevention (2014). TBI is often used to refer specifically to brain damage caused by hematoma, diffuse axonal injury, or concussion, but skull fracture occurrences are documented in this category as well. While the presence of a skull fracture does not always directly pose a risk for brain injury, the conditions leading to fractures do. Additionally, the pathophysiology of the fracture varies depending on the strike. Open fractures breach the immunological barriers around the brain, allowing for infection, while depressed skull fractures can apply pressure directly to the brain, tear bridging veins, and allow for hematoma formation.

Soft tissue TBI can occur at thresholds lower than those needed to cause skull fracture, but the nature of the fracture can reliably indicate for the presence and severity of any soft tissue damage,

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http://www.techno-press.org/?journal=bme&subpage=7 ISSN: 2465-9835 (Print), 2465-9959 (Online)

as found in multiple experiments, such as those by Annegers and Coan (2000), Bazarian *et al.* (2005), Bruns and Hauser (2003). While linear skull fractures by themselves do not tend to complicate injuries, depressed fractures can physically press upon the brain, causing further damage (Hofman *et al.* 2000). The exact variety of fracture depends on the intensity, location, and other factors relating to the blow. Direct bending of the bony structures of the skull in high energy collisions can result in fracture by strains directly resulting from the impact, but linear fractures are also able to develop outside of the primary strike area due to the elastic nature of bone tissue and outbending that develops secondary to the strike (Gurdjian *et al.* 1950).

1.1 Causes of TBI

Likely TBI causes vary based on a subjects age and employment. Common civilian causes include falls, motor vehicle accidents, assault, and sports. Fall related fracture is the most common form of TBI and is especially prevalent in youths or the elderly, with an estimated one in three people over the age of 65 experiencing at least one fall per year (Watson and Mitchell 2011). For soldiers in combat, improvised explosive devices produce the majority of head injury (Galarneau *et al.* 2008), brought about by the pressure wave and shrapnel accompanying them (Holcomb *et al.* 2006). Due to the surroundings of each injury circumstance, each of these injury mechanisms has slightly different presentations of injury.

1.1.1 Causes of TBI - Falls

Falls are the most common cause of TBI, accounting for over a third of all cases (Cormier *et al.* 2011a). Although falls account for a lower percentage of TBI-induced fatalities compared to other modalities, the energy release from a short fall is still sufficient to cause skull fracture (Gennarelli *et al.* 1994). Fall induced TBI death rate for the elderly has been constantly increasing since the 1980’s (Watson and Mitchell 2011) despite an increase in the self-reported average health of the age group (Stevens and Adekoya 2001). Fall injuries most often appear in either single or multiple impacts around the “hat brim area” (Kremer *et al.* 2008), a 3 cm thickness region around the head with a lower limit formed from the circle connecting the top of the eyebrows to the occipital pole. TBI caused by assault can also present with similar symptoms to that caused by falls (Billmire and Myers 1985).

1.1.2 Causes of TBI - Motor Vehicular Collisions

Compared to falls, motor vehicle collisions can achieve much higher impact energies and cause more fatalities (Faul 2010). Due to advances in safety devices including airbags, many injuries are at least partially mitigated, however brain injury and skull fracture can and do occur. Frontal and rear impacts are less associated with soft tissue trauma when compared to lateral impacts, both due to the concentration of safety measures in cars to mitigate frontal impacts, as well as a lower innate resistance of the brain to lateral impact, as found in multiple studies including Hodgson *et al.* (1983), Nahum *et al.* (1968), Schneider and Nauhm (1972).

In the event of airbag failure, strikes to the steering wheel can cause fracture as well, even at relatively low velocities (Yoganandan *et al.* 1991). Similarly, side impacts can evoke head trauma when the head strikes the window, which is exacerbated by the smaller crumple zone in that direction. Side airbags help prevent injury in this direction, as do their frontal counterpart, but are not standard for all vehicles.

1.1.3 Causes of TBI - Sporting Impacts

Sports such as football and soccer are troubled with head impacts. National Collegiate Athletic Association (NCAA) defensive line football players receive up to 1440 head impacts per year (Forbes *et al.* 2010), with about 280 of the hits to the more sensitive lateral areas of the brain (Crisco *et al.* 2010). Rotational acceleration of the brain with respect to the skull are the major cause of many forms of TBI, and modern helmets are unable to reduce the propagation of all of the accelerations and forces associated with these impacts to non-injurious levels (Viano *et al.* 2006). Since 1945, impacts have resulted in over 350 American football player deaths due to subdural hematomas alone (Forbes *et al.* 2010).

1.1.4 Causes of TBI - Explosive Injury

Blast related TBI, found most commonly in improvised explosive device related injuries, is perhaps the most complicated head injury format. Primary blast neurotrauma is produced by the pressure wave formation of an explosive device. This traveling pressure wave is amplified while traveling under the helmet (Ling *et al.* 2009) and causes ripples in the skull that further amplify the damage done by the wave (Moss *et al.* 2009, Panzer *et al.* 2012). In conjunction to this, explosive devices almost always create a cloud of shrapnel, either by design of the device, or as a result of loose debris in the blast. This shrapnel creates secondary trauma to the body, as well as the possibility of penetrating head injury. In a study of 63 US military personnel with TBI ranging from mild to severe as a result of IED devices, not a single one presented with blast injuries that were not complicated with additional sources of injury (Mac Donald *et al.* 2011). Because of this, it is very hard to accurately model a realistic blast injury as no model can accurately predict the secondary injuries caused by an explosive event. Similarly, autopsy reports of those that have died from assault are often unable to determine if the injuries caused in these instances are due to the initial blows, or secondary injuries such as falls and other aspects of an assault (Graham *et al.* 1992).

1.2 Models of TBI

In an effort to understand TBI and better know how injuries form and how to prevent them, it is important to construct models of the head. Because of the obvious ethical issues involved, the only living human experimentation protocols for TBI must be well below levels that could cause injury to the patient. Other sources of information can be found in testing human cadavers, which allows for testing at supra-injury levels, however the supply of cadavers available for testing is limited. Animal testing allows for more extensive testing, however ethical issues still exist and the results are not directly transferable to human injury levels. Finite element modeling has the advantage of allowing for unlimited testing, however the accuracy of the results is limited by the power of the computer, the resolution of the structures being modeled, the accuracy of the imputed parameters, and the lack of variation of structure that would be found between different people. Therefore, examining results from each model type and comparing them is needed to construct the full picture of the sensitivity of the human brain.

Developed as a result of injury modeling, predictive equations are also used to attempt to indicate injury with the application of different forces and accelerations. The most widely used rating systems for head injury, which are also summarized in Table 1, are the Gadd Severity Index, the Head Injury Criteria, the Head Impact Power, the Linear Skull Fracture Criteria, the Simulated Injury Monitor (SIMon), and the Louis Pasteur University Model (ULP). Of these, the HIC and

Table 1 The most common head injury measurements and their limitations

Injury Scale	Formula	Measures	Limitations
Gadd Severity Index Gadd (1966)	$\int_{t_1}^{t_2} a(t)dt^{2.5}$	linear acceleration injury risk; t_1, t_2 = start and end points of measurement window	Same as HIC, but does not show the maximum index between two time points.
Head Injury Criterion Versace (1971)	$\left\{\left(\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right)^{2.5} (t_2 - t_1)\right\}_{\max}$	linear acceleration injury risk given a variable acceleration	Used mainly in crash testing. Uses acceleration from center of mass, does not take into account forces such as rotational acceleration, which is highly correlated to concussion.
Head Impact Power Newman *et al.* (2000)	$C_1 a_x \int a_x dt + C_2 a_y \int a_y dt + C_3 a_z \int a_z dt$ $+C_4 \alpha_x \int \alpha_x dt + C_5 \alpha_y \int \alpha_y dt + C_6 \alpha_z \int \alpha_z dt$	linear and rotational injury risk; C_n = risk weighting coefficients for acceleration vectors	Expansion of HIC. Little use for hematoma or fracture
Linear Skull Fracture Criterion Vander Vorst *et al.* (2003)	$\ln\left(\frac{P}{1-P}\right) = C_1 * \ln(strain) - C_2$	Skull fracture risk given g-force; P = fracture risk, C_n = constants	Does not distinguish between different impact conditions
Simulated Injury Monitor Takhounts *et al.* (2003)	FE model	Strain, dilation, and relative motion damage	Uses a rigid skull with low resolution. Low accuracy
Louis Pasteur University Model (ULP) Willinger *et al.* (1999)	FE model	Von Mises in the brain, strain energy leading to SDH and skull fracture	Low resolution finite element model using acceleration fields.
Linear Skull Impactor Model (LSIM)*	F= $5534 + 417 * Weight - 369 * Velocity - 328 * Location$	Force in newtons for impactors of varying weight (in kg), velocity (in m/s) for the frontal and lateral sections of the skull	Linear nature of the equation provides an estimate of the magnitude of force, but does not fully capture experimental data.
Expanded Skull Impactor Model* *Models derived in this paper	Fracture Force=13562+1303*H1 -9703.7*H2-3057*H3 H terms are functions of the hyperbolic tangent and are omitted for brevity. See equation 2 for expanded form.	A more detailed version of the LSIM using a neural network model over linear equations.	Provides better accuracy than LSIM due to the inclusion of non-linear terms, but still loses accuracy as inputs deviate from the initialization parameters.

HIP are equations that take into account accelerative forces on the center of mass of a subject but fail to adequately describe several conditions of brain injury including those where the forces do not act about the center of mass. Additionally, they are of very little use for describing skull fracture. The SIMon and ULP are low resolution finite element models of the head which require acceleration inputs and a subsequent lengthy calculation time. The Linear Skull Fracture Criterion shows promise as a predictive statistic, but fails to account for the complex details of a collision like location and impactor shape.

All of the equations for determining injury score are mired in controversy. Though it is the standard for automotive safety testing, experts disagree over what thresholds represent a suitable tolerance limit as well as the maximum allowable pulse length for the HIC. Other papers conclude that the very foundations of the HIC formula are invalid, with other factors besides linear acceleration being the main contributor to injury (Newman 1980). Other studies have shown very little correlation between the HIC predictor of injury and in-hospital ratings of injury such as the Abbreviated Injury Scale (Ommaya 1981). Compounding to this is the variability of physiology between ages and sexes, not to mention the variability from person to person within these groups. This variation forces equations like the HIC and the HIP to use a generalized model of the skull and make large assumptions. Additionally, these formulae fail to distinguish between mechanical and biological damage to tissue, which can be important as biological disruption in messaging between neurons can be disrupted long before mechanical failure occurs. Therefore, new models are required that push the forefront of head injury knowledge to accurately determine the likelihood of head injury in various scenarios. This need gave the impetus to derive a new predictor for head injury, based on the parameters seen in head impact testing.

2. Data collection

As can be expected, the more energetic the impact, the more likely it is for that bone to suffer a mechanical failure. The speed of the impactor, the weight with which it impacts, and the location of the strike are all variables in determining the failure point of the bone. Due to the limited availability of human skulls for testing purposes and the limited amount of data that can be acquired from each one, there is a gap in literature of exact values for skull fracture. Nevertheless, data was drawn from several sources to form an analysis of the parameters needed to cause fracture. A brief description of these studies follows, along with a summary of the reported range of forces at failure, which can be seen in Fig. 1. The range of values for each column shows the drastic variation in fracture force caused by the different input parameters.

In a study in 2011 by Cormier *et al.*, using a free-fall impactor, acoustic sensors were attached to the skull to detect the breaking point for the frontal bone. The skulls were rigidly attached at the occipital lobe to prevent movement, and the flat surface of a 6.45 cm^2, 3.2 kg cylindrical impactor was allowed to fall onto the frontal bone. Analysis of the precise area of impact by means of film in the impacting area allowed for precise calculation of the force experienced by the bone. The study found a 50% risk value for frontal bone fracture at forces for this impactor at values between 1885 and 2405 N, although it was noted that the natural variation between the frontal sinus cavities of patients affected the fracture sensitivity (Cormier *et al.* 2011). A similar study by Nahum in 1968 used a 2.9 cm diameter impactor and showed fracture force values between 4050 and 6300 N for frontal impact and 3050 to 3980 N for side impact (Nahum *et al.* 1968). In another study by Allsop in 1992 on the frontal bone, the impacting areas consisted of a flat plate and a 6.45 cm^2

Fig. 1 Chart of selected bone failure values based on position and study 1) Cormier *et al.* (2011b), 2) Swearingen (1965), 3) Yoganandan *et al.* (1991), Yoganandan *et al.* (1995), Yoganandan and Pintar (2004), 4) Hodgson (1967), Hodgson *et al.* (1970), Hodgson *et al.* (1983), 5) Nahum *et al.* (1968), 6) Schneider and Nahum (1972), 7) Allsop *et al.* (1992)

cylinder with velocities of 4.3 m/s and 2.7 m/s respectively. The mean fracture forces were 12390 and 5195 N respectively (Allsop *et al.* 1992).

Access to devices such as the ones used in the aforementioned studies, which allowed for precise measurement of force values, velocities, and other details about the injury, are uncommon. In one study, Swearingen determined the forces in a car accident by purchasing dash panels with indentations in them from the passenger's heads impacting with the dash from cars that had been involved in frontal collisions. They then put a model head in a catapult and adjusted the force values until they got the same dent parameters in new dash panels. As a follow-up, they examined the fracture tolerance of various bones using a 6.45 cm^2 impactor placed on the catapult arm. Fracture values of 120 and 180 g's were found for the frontal bone, and they noted that increasing the area of impact increased the fracture tolerance beyond the force generation capabilities of their device (Swearingen 1965).

Similar to a free-fall impactor scenario, head drop experiments such as the ones carried out by Hodgson in a series of experiments swap velocity vectors such that it is the impacting surface that is stationary and the body that moves. Impacts by this means were carried by multiple drops for each head, ranging from about 0.1 to 1.1 meters onto probes mounted on a load cell. The experimenters used profiles of various shapes including plates, 6.45 cm^2 cylinders, and hemispherical anvils. The actual area of impact was not measured in these experiments although the head was restrained by a cord to maintain the proper striking location during fall. Additionally, because fracture does not always occur directly at the impact site, it was not possible to give exact geometric coordinates for the impact. Forces ranged from 3114 to 7340 N of force in fracture scenarios in Hodgson (1967); Hodgson *et al.* (1970); Hodgson *et al.* (1983).

Other studies used a pneumatic striker instead of a free fall scenario. While this allows for precise velocity control and measurement of force, it is not mimetic to the events in an accident, where ballistic laws are in effect. In several studies lead by Yoganandan, measurement of force values was conducted at quasistatic (2.5 mm/s) and dynamic loading (7.1 to 8.0 m/s) with a 9.6 diameter hemispherical impactor. Since the actual area of impact was not measured in the study, the impact area cannot be said with certainty, however force values for the frontal bone ranging between 4642 and 13600 N. Likewise values for the temporoparietal bones range from 5603 to 612 N in Yoganandan *et al.* (1995), Yoganandan and Pintar (2004).

3. Results and discussion

Since studies rarely show exact geometric coordinates for the impact areas on the skull, examples of the fracture tolerance force are shown based on the general classification of the frontal bone or the side (temporoparietal). The tolerance values from the previously listed studies were compiled in the JMP statistical software package and analyzed for trends to derive a model. Based on this analysis, a simple, multivariate equation, shown in Eq. (1) was developed by means of computational least squares analysis for the fracture force

$$\text{F}= 5534 + 417 * Weight - 369 * Velocity - 328 * Location \tag{1}$$

Eq. (1) was achieved with an R^2 of 0.412 where location was 1 for the frontal bone and 2 for the temporoparietal area. In addition to the exclusion of the precise impact area, the area of the impactor was not able to be included in the equation as most studies investigated in this paper described the isolated surface of the impactor, but did not give an accurate representation of the area of the impactor involved during a specific impact testing, see Allsop *et al.* (1992), Hodgson (1967), Hodgson *et al.* (1970), Hodgson *et al.* (1983), Nahum *et al.* (1968), Schneider and Nauhm (1972), Yoganandan *et al.* (1995), Yoganandan and Pintar (2004). Therefore, no significant relationship between the area of the impactor and the striking force was found.

The low R^2 value found in equation 1 suggests a high degree of nonlinearity, implying that Eq. (1) does not fully describe the data represented here, and was therefore refined to an R^2 value of 0.978 using a neural model of the data generated by the JMP software package, as seen in Fig. 2. The neural network uses the series of formulas constituting Eq. (2). As with the initial model, location was a binary variable with 1 representing impacts to the frontal bone and 2 representing impacts to the temporoparietal area, weight is in units of kg, and velocity is in units of m/s

$$\boldsymbol{Fracture\ Force = 13562 + 1303 * H1 - 9703.7 * H2 - 3057 * H3}$$

$$\begin{aligned} H1 &= TanH\big(0.5(-8.2 + 0.561 * Weight + 0.820 * Velocity + 0.554 * Location)\big) \\ H2 &= TanH\big(0.5(-4.61 + 0.164 * Weight + 0.822 * Velocity + 1.894 * Location)\big) \\ H3 &= TanH(0.5(7.20 + 0.022 * Weight - 0.0776 * Velocity - 3.516 * Location)) \end{aligned} \tag{2}$$

Many current finite element models treat the skull as a purely elastic construct. Such modeling neglects the study of impacting speed and its relationship to the breaking point of materials. Based on the results derived from this equation, the non-zeroth-order nature of the skull with respect to velocity is clearly shown by the change in fracture force. Furthermore, the exact decrease in force tolerance with increasing energies due to both the mass and the velocity of the impacting object are shown in Fig. 3.

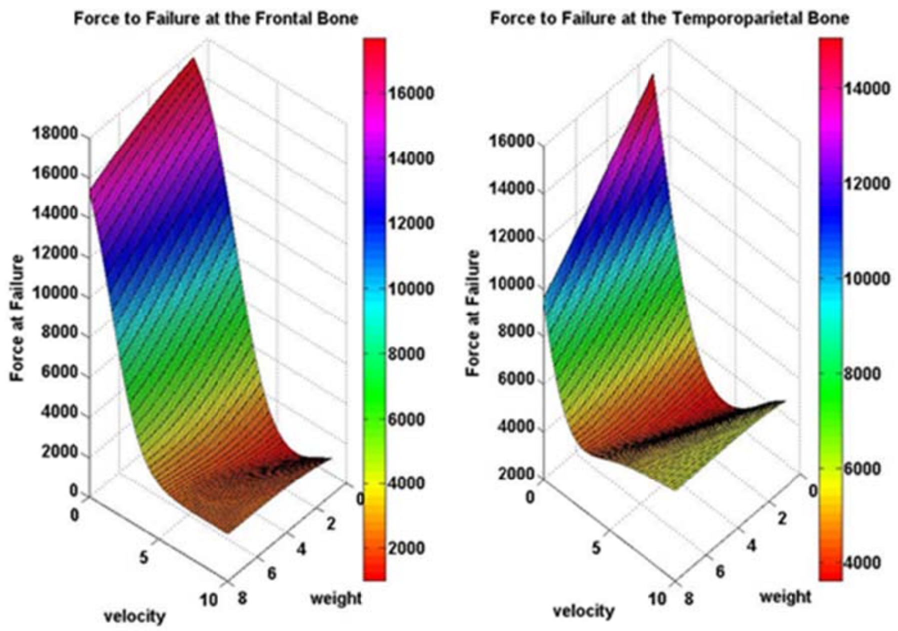

Fig. 2 Predicted force required to cause fracture. Left=Frontal, Right=Temporoparietal

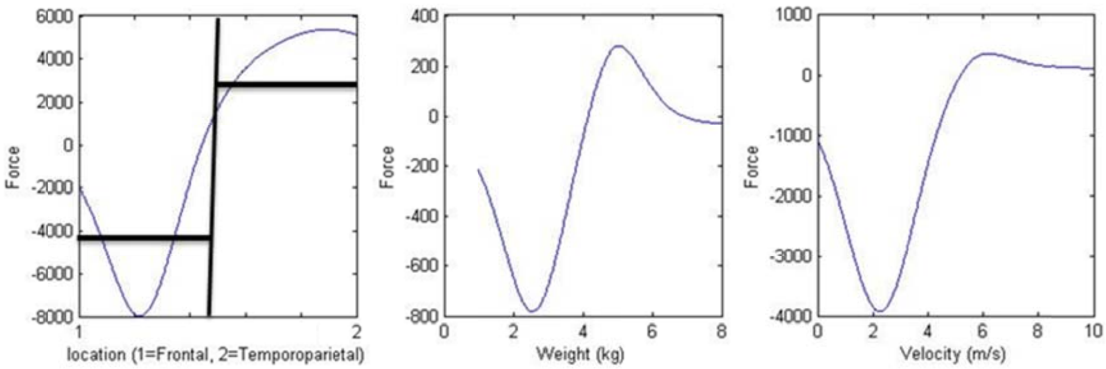

Fig. 3 Sensitivity profiles of (top to bottom) velocity, weight, and force with respect to fracture force.

Though it is known that side impacts have a higher risk for concussion and other forms of brain damage when compared to frontal collision, several studies have shown conflicting values for the risk functions of skull fracture. The equation found here shows an average breaking force to be 3 kN higher for the frontal portion of the skull, implying that the temporoparietal bone is weaker to impact, similarly to the brain. This relationship does not hold static across all weights and velocities however. When impactor velocity was held near its upper studied bound, the dynamic was reversed, with frontal impacts becoming more sensitive (~2000 N) when compared to side impacts (~6000 N) for similar weight impactors. The equivalence point where both areas had an equal risk for the average impactor weight used here was 4.10 m/s.

Points where the sensitivity analysis detected no change in the breaking force were also found.

These existed as a maximum at 5.6 m/s for velocity and as a maximum and minimum at 4.15 and 6.93 kg for the mass. Additionally, velocity rising and falling inflection points - the locations of the most change in force per unit velocity - were found at 2.50 (derivative value of -3936 N/(m/s)) and 4.96 m/s (359 N/(m/s)) respectively. Mass rising and falling inflection points were found at 2.25 (derivative value of -785 N/kg) and 6.24 kg (277 N/kg) respectively. Although the sensitivity of force due to mass shows an apparent absolute gain in amplitude by passing though the zero point at the end of the data range, both mass and velocity sensitivity profiles show low absolute values for the rate of change in fracture sensitivity at this region, suggesting that the fracture tolerance of bone changes only marginally with proportionally larger changes in mass and velocity.

4. Conclusions

In order to more accurately understand skull fracture and predict when it will occur, this study compiles and analyzes the results of past fracture experiments to form a novel predictive equation for head impact based on the velocity, mass, and location of a striking object. This novel equation shows that the sensitivity of the skull to impact varies considerably based on the location and energies of the impactor, with an average of the frontal bone resisting 3 kN greater force than the temporoparietal bones. When velocity was fluctuated within the range of prior experimental testing, the lateral side was found to be more sensitive in low velocity impacts and the frontal side more sensitive at high velocities. While the neural network is not accurate outside of its data range, this research provides a relatively simple equation for the calculation of head injury risk for smaller diameter impactors. This model was created with a high R^2 value against the point data from accepted studies, which helps to fill the gaps in our knowledge by analyzing the data currently available and guiding head injury research in the future.

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