Incorporating Human Body Mass in Standards of Helmet Impact Protection against Traumatic Brain Injury

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ABSTRACT

Impact induced traumatic brain injury (ITBI) describes brain injury from head impact not necessarily accompanied by skull fracture. For sufficiently abrupt head impact decelerations, ITBI results from brain tissue stress incurred as the brain crashes into the inside of the skull wall, displacing the surrounding cerebral spinal fluid (CSF). Proper helmet cushioning can damp the impact force and reduce ITBI. But force is mass times acceleration and current helmet blunt impact standards are based only on acceleration thresholds. Here I show how this implies that present standards grossly overestimate the minimum acceleration onset for ITBI by implicitly assuming that the brain is mechanically decoupled from the body. I quantify how an arbitrary orientation of the body with respect to impact direction increases the effective mass that should be used in calculating the required damping force and injury threshold accelerations. I suggest a practical method to incorporate the body mass and impact angle into ITBI helmet standards and point out directions for further work.

1. Introduction

Traumatic Brain Injury (TBI) refers to physical injury to the brain not necessarily accompanied by visible external head injury (e.g. Bandak et al. 1996). Impact induced TBI (ITBI) arises from rapid acceleration or deceleration of the head, as can occur in sports, motor vehicle accidents, or in military combat. The brain is composed of soft tissue and is surrounded by a layer of cerebral spinal fluid (CSF) inside the head. During normal head motions, the force on the head is small enough that the CSF prevents the head from impacting the skull wall. In contrast, the magnitude of deceleration upon impact is so large that the CSF cannot adequately protect the brain. As the skull comes to a stop, the brain pummels the the inner skull pushing the fluid away. The brain deformation can occur fast enough to leave a small cavity between skull and brain at the antipode. As the brain rebounds it slaps into the CSF such that countre-coup injury can occur. The brain deformation can also induce more diffuse brain tissue injury via shear stress.

The medical consequences of such ITBI range from minor concussions to complete cognitive impairment. Almost 2 million civilian cases of TBI (virtually all are ITBI) have been diagnosed each year since the early 1990s, leading to 200 hospitalizations per 100,000 people each year and 56,000
deaths per year (McArthur et al. 2004). About 50% of TBI cases come from automobile accidents and 20% from sports related injuries (Bohnen et al. 1992). The majority of TBI are classified as mild traumatic brain injury (MTBI) or concussion. TBI is also common among military combat personnel (Okie 2005). Recent estimates suggest as high as \( \sim 20\% \) of US soldiers returning from Iraq and Afghanistan have incurred TBI (Terrio et al. 2009). In this context, TBI is likely a combination of ITBI and blast induced TBI (BTBI), the latter referring to the direct effect of blast overpressure (Cernak 2005; Taber et al. 2006; Moss et al. 2009) which adds to any additional ITBI.

Protection against ITBI requires helmets with proper cushioning and a proper blunt impact measure to determine the effectiveness of such helmets. Current helmet blunt impact standards are derived from empirically determined injury measures of acceleration vs. duration based on cadaver and scaled monkey data (Ono et al. 1980). Drop tests of helmeted head forms fitted with accelerometers for chosen drop heights then empirically test whether a given helmet falls within the acceptable acceleration range upon impact (e.g. McEntire et al. 2005). But force equals mass times acceleration, so the use of acceleration thresholds without incorporating head and body mass is flawed. Using only the head form + helmet mass may be appropriate for computing the impact force on the head for a body oriented perpendicular to the direction of impact, but the effective mass increases for impacts with the body increasingly aligned with the direction of impact because some fraction of the force incurred by the body is transmitted through the skull to the brain. For exact alignment, the force would depend on the entire body mass.

In section 2, I give a simple derivation of the physics principles behind helmet protection to blunt impact and ITBI. In section 3, I discuss the quantitative flaws of current blunt impact/TBI standards. In section 4, I derive corrections to standard threshold TBI measures that incorporates the body impact angle and thus the effective mass of impact. In section 5 I describe how these corrections can be implemented in practice in future work and conclude in Sec 6.

2. Why Cushioning Reduces Impact Force

Newton’s equation of motion for an object of mass \( m \) subject to a force is

\[
\frac{dv}{dt} = F = ma, \tag{1}
\]

where \( F \) is the force, and \( v \) is the speed, and \( a \) is the acceleration. For an object incurring a drop and impact, Eq. (1) is used to compute the motion during free fall, and the deceleration upon impact determined by the helmet properties. Using \( a = (0, 0, a) \) and \( v = (0, 0, v) \) (i.e. both with only \( z \) components) and assuming that \( |\frac{dv}{dt}| \ll |\frac{da}{dt}| \) we can integrate (1) equation to obtain

\[
v(t) = v_0 + at, \tag{2}
\]

where \( v_0 \) is the initial speed at initial height \( z_0 \). Integrating (2) gives

\[
z(t) = z_0 + v_0 t + \frac{1}{2} at^2. \tag{3}
\]
Eliminating $t$ from (2) and (3) gives
\[ v^2(t) - v_0^2 = 2ad, \]
where $d = z(t) - z_0$.

We can use (4) to compute the maximum free fall speed reached just before impact for an object dropped from rest at height $h$ when $a$ is given by Earth’s gravitational acceleration $g = -10 \text{m/s}^2 = -32 \text{ft/s}^2$. For $v_0 = 0$, $z_0 = h$, and $z(t_I) = 0$, Eq. (4) implies the speed toward the ground at the time of impact $t_I$ is
\[ v_I = v(t_I) = (2gh)^{1/2}. \]

This gain in speed corresponds to gain kinetic energy at the expense of gravitational potential energy. Upon impact, most of this kinetic energy is converted into work done in deforming and stopping the object. This work can be expressed as the force incurred times the stopping distance, and equals the kinetic energy just before impact. That is,
\[ F_s s = \frac{1}{2}mv_I^2, \]
where $F_s = ma_s$ is the force exerted on the object by the stopping acceleration $a_s$ upon impact over the stopping distance $s$. Combining Eq. (5) and Eq. (6) gives
\[ a_s = \frac{v_I^2}{2s} = \frac{gh}{s}, \]
showing that increasing $s$ reduces the magnitude of acceleration and thus force of impact (see also Cory et al. 2002).

A larger stopping distance $s$, also implies a longer stopping time: By applying equation (4) to the case in which the object’s initial speed corresponds to the speed of impact $v_I$ from (5) and taking the final speed $v(t > t_I) = 0$ as the object comes to rest, we obtain $v_0 = (2gh)^{1/2}$. Plugging this into (3), setting $|z(t) - z_0| = s$, and using (7) we have
\[ s = (2gh)^{1/2}t + \frac{gh}{2s}t^2. \]
Solving (9) for $t > 0$ gives
\[ t = (2 - \sqrt{2})\frac{s}{(gh)^{1/2}} \]
which highlights that the longer the stopping distance, $s$, the longer the deceleration time $t$ for an object that acquired its impact speed by falling from height $h$.

Eqs. (7) and (9) show that if cushioning can increasing the distance or time over which a headform decelerates from its maximum speed to zero, the magnitude of acceleration of the impacting object is reduced, and thus so is the force of impact.
The amount of tissue damage and TBI depends on a combination of the external force and the time scale over which the force acts. Below some minimum threshold force, determined by the biological tissue properties, no damage will occur no matter how long the force is applied. However, a small force above this threshold acting over a long time could do more damage that a much larger force over a short time. For a given mass of impactor, empirically determined damage curves, in principle, provide a practical method for identifying an injury threshold curve in the force vs. duration plane. As I describe in the next section, present curves are constructed in the acceleration vs. duration plane, and practical application of these curves has fundamental shortcomings.

3. Shortcomings of Current Head Impact TBI Protection Indices

The peak acceleration incurred for a fixed mass impactor indicates the peak force providing one measure of potential injury. However, the need to incorporate a combination of acceleration and duration into an injury measure (see Hayes et al. 2007 for review) was evident from the Wayne State Tolerance Curve (WTSC) (Pattrick et al. 1963, Snyder 1970) supposedly be tolerated without severe head injury (considered to be skull fracture). The original data came from (1) drop tests of 4 embalmed cadaver heads on plates, with measurements of linear acceleration, intracranial pressure and skull damage (2) air blasts to exposed cadaver brains and (3) hammer blows to animals. The data showed that small accelerations can be tolerated for longer durations than large accelerations. The severity index (SI) (Gadd 1966) quantifies the WSTC into a (unfortunately dimensional) quantity given by

$$SI \equiv \int_{t_1}^{t_2} a_g(t)^{5/2} dt$$

where $a_g$ is the dimensionless acceleration in units of gravity $g$ and $t$ is measured in seconds.

The SI incorrectly implies that impacts of extremely slow deceleration extended over a very long period give the same injury threshold as high deceleration impacts of very short duration, whereas there is no injury at very low accelerations. This is partly corrected by the Head Injury Criterion (HIC) (Versace 1971)

$$HIC = \left[ \frac{(t_2 - t_1)}{t_2 - t_1} \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a_g dt \right\}^{5/2} \right]_{\text{max}},$$

which restricts the SI to an integral near an empirically chosen time interval (measured in seconds) $t_2 - t_1$ near the peak acceleration. When the acceleration magnitude is nearly constant over the chosen time interval, $HIC \approx SI \propto a_g^{2.5}$. Typically an HIC between 500 and 2500 is converted into a probability for fatality or concussion. As applied to non-fatal TBI, Ono et al. (1980) performed experiments both with human cadavers and live monkeys and determined distinct human thresholds of skull fracture vs. concussion, a form of TBI. The latter TBI threshold curve has been called the Japanese Head Tolerance Curve (JHTC).
But the SI and HIC standards are flawed. Note for example, that concussions in the NFL are occurring with higher probability that the JHTC curve would predict at measured values of the head acceleration (Viano et al. 2006). In addition, King et al. (2000, 2003) and Zhang et al. (2004) used video footage of helmet-helmet collisions in games from the National Football League (NFL) in which known concussions occurred. The motions producing the concussions were reproduced in the laboratory using helmeted dummies, with linear and rotational accelerometers. The data measured were then fed as initial conditions into head impact computer simulations that include a comprehensive computer model of the human head and brain (Wayne Stead Head Model). By analyzing the stresses on the simulated brain tissue, the HIC proved to be no better than the peak linear acceleration, or the head impact power (HIP, Newman et al. 2000), an uncommonly used measure of the total kinetic energy per unit time. The HIP was marginally the best correlator, followed by the peak acceleration and then the HIC, and rotational acceleration.

In principle, the conceptual advantage of the HIP would be that it includes mass whereas the SI, HIC and peak accelerations do not include the mass. However, typically the mass is used is that for the head itself not adjusted for impact angle and body mass. Also impacts analyzed from drop tests in the laboratory use approximately the same mass of head forms and helmets, so the relative change in effective mass as a function of body impact angle is not present. The non-inclusion of the mass is a conceptual shortcoming that I quantify in the next section.

4. Incorporating Impact Area and Body Mass into TBI Protection Standards

Mechanical stress on brain tissue causes TBI and how external impact forces produce specific clinical manifestations of ITBI comprises a complex set of questions. But the role of helmet protection is largely independent of the specific TBI manifestation: a helmet accomplishes much by simply reducing the overall stress on brain tissue. For a given acceleration and fixed mass, the local stress is reduced for a larger brain surface area of impact. For a given surface area and a given acceleration, an increased mass will produce more force per unit area and thus more stress.

Using the reasonable assumption that that material threshold for brain tissue damage is the same in woodpeckers and humans, Gibson (2006) showed why woodpeckers would not be expected to get concussions even though they incur high enough accelerations over long enough durations to exceed the TBI threshold of the JHTC curve. The material stress associated with the very high HIC value is still below the tissue damage threshold when applied to a woodpecker head. The same HIC value corresponds to a much higher force per unit area when applied to the human head.

Complementarily, when comparing impacts of the same surface area but different effective masses, a single HIC standard is also inadequate because force per unit area depends on mass. A person oriented vertically during a fall on their head will incur more head force compared to a person oriented horizontally during the fall. The stress incurred by the brain as it contacts the inner skull during the head impact is a combination of (1) the force need to stop the brain as if
it were isolated in free fall, plus (2) a contribution that comes from stress waves propagating into the brain from the skull which are sourced by the weight of the entire body which is coupled to the brain via the CSF. The force on the skull depends on the mass aligned along the direction of impact, but because the coupling between the skull and brain is likely less than 100% efficient, there is some efficiency coefficient that scales the force on the skull to that on the brain for a fixed effective mass along the direction of impact. In a more detailed study, it may turn out that this coefficient depends on mass but here I consider the simple case in which it does not, and then take ratios of quantities in which the coefficient cancels out.

To see the role of the body mass quantitatively, let \( 0 \leq \theta \leq \pi/2 \) be the impact angle between the line passing through the body center of mass and impact point, and the line through the impact point in the direction of center of mass momentum before impact. The case \( \theta = \pi/2 \) corresponds to the body oriented horizontally for a vertical fall impact and \( \theta = 0 \) corresponds to the body oriented vertically for a vertical impact. For any \( 0 \leq \theta < \pi/2 \), the effective mass of the head will be larger than just that of the head form.

Consider two cases labeled by 1 and 2, which respectively produce a force per unit area of \( \sigma_1 \) and \( \sigma_2 \) on the brain. Let \( \sigma_c \) be a property of brain tissue indicating the threshold stress above which TBI occurs. Taking \( \sigma_1 = \sigma_c = \sigma_2 \), and expressing \( \sigma_1/\sigma_2 \), in terms of the separate properties of each system, we have

\[
\frac{\sigma_1}{\sigma_2} = 1 = \frac{F_1A_1}{F_2A_2} = \frac{[(m_{b1} - m_{h1})\cos\theta_1 + m_{h1}]a_1A_1}{[(m_{b2} - m_{h2})\cos\theta_2 + m_{h2}]a_2A_2}, \tag{12}
\]

where \( F_1 \) and \( F_2 \) are the forces on the respective heads during impact deceleration; \( A_1 \) and \( A_2 \) are the head contact areas; \( m_{h1}, m_{h2} \) and \( m_{b1}, m_{b2} \) are the respective head and total body masses for the two cases, \( \theta_1, \theta_2 \) are the respective impact angles, and \( a_1, a_2 \) are the magnitudes of the deceleration from maximum to zero upon impact. If the two impacting bodies are identical but differ in impact angles for the cases considered, we can set \( A_1 = A_2, m_{h1} = m_{h2} = m_h, m_{b1} = m_{b2} = m_b \) in (12). After a bit of algebra, this gives

\[
\frac{a_1}{a_2} = \frac{(m_b/m_h - 1)\cos\theta_2 + 1}{(m_b/m_h - 1)\cos\theta_1 + 1}. \tag{13}
\]

If we take \( \theta_2 = \pi/2 \) as a fiducial baseline case corresponding to the body perpendicular to the direction of impact, we obtain

\[
\frac{a_1}{a_2} = \frac{1}{(m_b/m_h - 1)\cos\theta_1 + 1}. \tag{14}
\]

This formula is plotted in Fig. 1a. For a fixed head+helmet mass of \( m_h = 6.4 \text{kg} \), the three curves in the figure correspond to body masses of \( m_b = 64, 82, 100 \text{kg} \) respectively. Each point on these curves corresponds to the impact force imparting the same TBI threshold stress. The curves show that this stress arises for a significantly lower magnitude of head acceleration when the body angle of impact deviates from the fiducial angle of \( \pi/2 \).
For most arenas of injury (e.g., football, military, motor vehicle accidents) the angle of impact will vary from incident to incident so practical incorporation of the effect of angle into a TBI standard requires either a conservative standard that protects for impact angles down to a chosen minimum \( \theta_{\text{min}} \), or a suitable average over a range of angles. For the latter, the average of (14) in spherical polar coordinates is

\[
\left\langle \frac{a_1}{a_2} \right\rangle = \frac{\int_{\mu_{1,\text{min}}'}^{\mu_{1,\text{min}}'} \frac{1}{(m_b/m_h - 1)\mu_1' + 1} d\mu_1'}{\int_{0}^{\mu_{1,\text{min}}'} d\mu_1'},
\]

where \( \mu_{1,\text{min}}' = \cos \theta_{1,\text{min}}' \), the cosine of the minimum impact angle (where \( \cos \theta_1 = 0 \) corresponds to impact direction perpendicular to body alignment and \( \cos \theta_1 = 1 \) corresponds to body aligned parallel to direction of impact). Fig. 1b shows plot the average of (14) over \( \theta_{\text{min}} \leq \theta \leq \pi/2 \) as a function of the choice of \( \theta_{\text{min}} \).

As discussed in Sec. 2., for a nearly constant acceleration over impact duration, the HIC and the SI are proportional to \( a^{2.5} \). By analogy to (14) and (15) we can then write

\[
\frac{HIC_1}{HIC_2} \approx \frac{SI_1}{SI_2} \approx \frac{a_1^{2.5}}{a_2^{2.5}} = \frac{1}{(m_b/m_h - 1)\mu_1' + 1},
\]

and for the average

\[
\left\langle \frac{HIC_1}{HIC_2} \right\rangle \approx \left\langle \frac{a_1^{2.5}}{a_2^{2.5}} \right\rangle = \frac{\int_{\mu_{1,\text{min}}'}^{\mu_{1,\text{min}}'} \left( \frac{1}{(m_b/m_h - 1)\mu_1' + 1} \right)^{2.5} d\mu_1'}{\int_{0}^{\mu_{1,\text{min}}'} d\mu_1'}. \tag{17}
\]

Eqs. (16) and (17) are plotted in the bottom row of Fig. 1 for mass ratios \( m_b/m_h = 10, 12.81, 15.63 \) corresponding to the bottom, middle, and top curves respectively in each panel.

5. Prescription for Revising ITBI Helmet Standards

The shapes of all curves in Fig 1. flatten at small \( \theta \) and steepen near \( \theta = 1.1 \) (\( \sim 63 \) degrees). Thus for either row 1 (peak acceleration) or row 2 (HIC), there is is a dramatic drop in the critical thresholds for injury even for angles that deviate only \( \sim 30\% \) from the fiducial \( \theta = \pi/2 = 1.57 \) rad. Complementarily, the curves in the right column panels of Fig 1. highlight that if impact angles are quasi-random over a range, averaging over this range for different choices of minimum impact angle \( \theta_{\text{min}} \) is relatively insensitive to this choice for \( \theta_{\text{min}} < 1.1 \) rad. Overall, the plots show that it may not be too much more demanding to protect against the full range of impact angles below \( \theta < 1.1 \) rad than it is to protect impact angles \( 1.1 < \theta \leq \pi/2 \) rad.

Using the calculations and Fig. 1, a procedure for straightforward improvement of ITBI helmet protection standards emerges: (1) Identify either the peak acceleration, or the SI, or HIC index on the usual JHTC type curve corresponding to the supposed acceptable injury threshold for the
characteristic time scale characteristic of the particular impact (e.g. football helmet collision). This provides the fiducial acceleration, the value corresponding to $\theta = \pi/2$ in the calculations above. Assume that this threshold is correct to produce supercritical stress on brain tissue for TBI based on acceleration of only the head form (e.g. helmet + head). (2) Pick a standard head mass AND body mass for a standard victim based on a practical statistical criterion of characteristic individuals involved. (3) Choose a characteristic minimum impact angle to accommodate a statistically significant fraction of all impacts based on carefully assessment of the types of impacts occurred in the activity and the equivalent range of impact angles. (4) Find the correction factor to the peak acceleration or HIC compared helmet drop tests in the laboratory either from plots like the left panels of Fig. 1 for the chosen angle, or the right panels of Fig. 1. using the chosen angle as the lower bound for averaging over an angular range.

For example, consider a head impact duration to be $\sim 15\text{ms}$. The 30% risk for concussions using the conventional JHTC corresponds to $a_1 = 125g$ for this duration. Let us assume that this is the correct threshold for TBI based on acceleration incurred for a helmeted head form of $m_h \sim 6.4\text{kg}$ in a drop test. Now take $m_b = 100\text{kg}$ so that $m_b/m_h = 15.63$ and consider typical collisions to take place at an equivalent impact angle range between $\theta_{\text{min}} = \pi/4 \leq \theta \leq \pi/2$. Using the bottom curves in the plots on the right column panels of Fig. 1., (which correspond to the chosen mass ratio), we find a reduction to the peak acceleration index threshold by a factor $1/4$, and a reduction to the HIC threshold by a factor $1/20$.

6. Conclusion

Commonly used ITBI helmet protection standards are based on empirical injury threshold curves of acceleration vs. impact duration from motor vehicle crash studies. Presently, helmet blunt impact and ITBI protection testing typically evaluate whether the acceleration upon impact from drop tests of helmet-fitted head form falls sufficiently below the injury threshold from these curves. I discussed that the resulting curves from this procedure can significantly overestimate the minimum acceleration for ITBI because they do not take into account the body angle of impact, and thus the effective mass of impact. The force on the head and brain is mass times acceleration and standard measures of protection based on acceleration can at most apply to a fixed mass of impactor. This absence of inclusion of effective mass in standard blunt impact criteria such as peak acceleration or HIC may explain for example, why concussions are seen in NFL football at lower peak accelerations than expected.

By incorporating the body impact angle with a simple practical paradigm, I showed that current blunt impact ITBI protection standards which utilize drop tests to compare with peak acceleration or HIC apply only for a body impact angle of $\pi/2$ (a horizontal fall to the ground). A correction to include the effective mass of impact for a 25% deviation from this impact angle requires a factor of $\sim 4$ drop in the acceleration threshold for injury. The calculations also show however, that the correction need not much exceed this factor of 4 to accommodate almost the full
range of effective impact angles.

The presentation has been minimalist to illustrate the key ideas. More detailed development and application of these concepts is warranted as the practical payoffs for ITBI protection are likely to be substantial.

The HIC in its current form does not accommodate variations in surface area or effective mass. The correction for mass is most important as it varies substantially from body impact orientation and its absence may help explain why e.g. NFL football TBI injuries are found even for lower values of the HIC than expected based on the JHTC curve (Viano et al. 2006). The JHTC was based on moving plate impacts onto seated monkeys with bodies restrained. The threshold acceleration derived from these experiments may be appropriate when the body is supported as in a car crash, but applies to e.g. a football impact at most only for the case in which the effective mass is that of the head + helmet. The latter would be the case only when the body is strictly perpendicular to the direction of impact. In general, injury acceleration thresholds inferred from JHTC type data can dangerously exceed relevant injury thresholds.

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Fig. 1.— Left panels: Threshold acceleration and HIC for impact induced TBI as function of impact angle between body and impact direction normalized to thresholds for an impact angle of 90 degrees. Right panels: Threshold acceleration and HIC averaged over angle from $\theta = \theta_{\text{min}}$ to $\theta = \pi/2$ plotted vs. $\theta_{\text{min}}$. In each panel the three curves represent, from top to bottom, a ratio of body to head+helmet mass of $m_b/m_h = 12.8, 10, 15.6$ respectively. Larger body to head mass means a lower acceleration threshold for injury for all angles except $\pi/2$ for a fixed $m_h$. 
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