ENHANCED STATISTICAL ANALYSIS OF HEAD INJURY DATA

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Abstract

Despite advances in road safety, head injuries still account for many of the most serious and fatal injuries in road traffic accidents. As such, traumatic brain injury is an important global public health problem.

Over the past half a century, a wealth of research has been carried out in an effort to determine the principal mechanisms responsible for head injuries. However, there are still conflicting notions on the mechanisms that cause head injuries.

A possible explanation for this incoherence is the confounding factors that appear to have been explored insufficiently in prior research work. Therefore, this study analysed the previous head injury research in greater detail.

Published head injury research from the past 70 years has been reviewed to:

- Collate existing propositions with respect to mechanisms for head injuries
- Catalogue the criteria that could be used to assess head injury risk
- Draw together a dataset of head injury case data; including an assortment (630 cases) of impact types, impact directions, contact surfaces, species (human or monkey), etc.
    - o Compare this dataset with expected results based on previously published assertions
    - o Make use of statistical techniques for this analysis

Peak linear head acceleration and the Head Injury Criterion (HIC) were found to be reasonable predictors of serious (Abbreviated Injury Scale, AIS ≥ 3) head injury occurrence. However several features of the impact conditions were shown to contribute to the injury outcome. Injury thresholds should take account of the confounding factors such as the specific impact conditions.

Introduction

Despite advances in road safety, head injuries still account for many of the most serious and fatal injuries in road traffic accidents and in trauma in general (Tagliaferri *et al.*, 2006). Traumatic brain injury is an important global public health problem (Hyder *et al.*, 2007). According to Critchley and Memon (2009), traumatic brain injury is a major cause of disability and survivors can often suffer cognitive, mood, and behavioural disorders; the societal cost of which can be substantial due to loss of years of productive life and a need for long-term or lifelong services. In their review of secondary safety priorities, Welsh *et al.* (2006) identified head injuries as being the most costly for vehicle occupants, when minor injuries were excluded.

Over the past half a century, a wealth of research has been carried out in an effort to determine the principal mechanisms responsible for head injuries. However, there are still conflicting notions regarding the mechanisms that cause head injuries. It is considered that a possible explanation for this incoherence is the confounding factors that appear to have been explored insufficiently in prior research work. Therefore, this study analysed the previous head injury research in greater detail. The objective was to identify the confounding factors from previous research that contribute to conflicting notions regarding head injury mechanisms and inhibit the development of advanced head injury criteria. To start this process the literature was reviewed to establish the current position in head injury research with regard to:

- Previous assertions as to head injury mechanisms
- Methods of describing head motion using accelerations
- Existing criteria

A summary of the research reviewed in relation to each of these topics is given in the following sections along with some discussion.

Injury mechanisms

As King *et al.* (2003) set out, "the precise mechanisms of brain injury have not been fully established and methods of prevention cannot be fully effective if we do not know the cause." However, whilst this may be the case, it has not prevented many authors hypothesising as to the mechanisms behind head injuries.

Holbourn (1943) hypothesised that translational acceleration of the head would not produce significant deformations in the brain due to the incompressible nature of the confined brain tissue. Thus he concluded that shearing deformations, which produce no volume change, caused by rotational acceleration could develop the shear strains throughout the brain required to produce the diffuse effects needed for concussive brain injuries.

Ommaya and Gennarelli (1974) put forward the hypothesis that, "rotational components of accelerative trauma to the head produce a graded centripetal progression of diffuse cortical-subcortical disconnection phenomena which is always maximal at the periphery and enhanced at sites of structural inhomogeniety." They proposed that the translational components of such trauma would be significant in the production of focal injuries only.

Based on sagittal plane acceleration impulse loading to the head of rhesus monkeys, and consideration of the resulting incidence of brain contusions, Gennarelli *et al.* drew the following conclusion:

> *"In the case of the frontal lobes, the tangential component and in the case of the inferior temporal lobe, the radial component of the brain movement appear to be the injurious factors (Gennarelli et al., 1979)."*

Melvin and Lighthall (2002) state that injury will occur if the magnitudes of the deformations and stresses induced in the tissues are sufficiently great. Therefore in order to develop truly predictive injury criteria, tissue stresses and strains must be related to dysfunction in physiological processes. Unfortunately, the measurement of strain is almost impossible during an impact, particularly *in vivo* (King *et al.*, 2003).

Motion described through acceleration

There has been some discussion in the literature regarding the importance of head acceleration to the mechanism of injury. Conventionally, acceleration modes are described either in terms of a rotation or translation of the head. This division has been adopted in the head injury literature and has become something of a focal point for much discussion as to which is the key measure: linear (translational) or angular (rotational) acceleration. The consideration of head motion as either linear or rotational is, however, somewhat misleading as one can be easily defined in terms of the other.

It seems that one key difference between translational and rotational motion for the head is that rotation implies varying tangential acceleration along the length of the radius. In pure translational motion all parts of the head will accelerate with the same magnitude (if a solid body). This is likely to be of great importance when considering the occurrence of intracerebral injuries. Intrinsically rotational motion of the head (if assumed to be solid/homogeneous) will tend to induce more shear distortion in the brain than pure translation (which without edge effects would induce no shearing within the bulk of the head). It seems that this implicit difference between the two descriptions of motion is rarely noted, but could be quite fundamental to our understanding of head injuries.

As we know that the head is not a solid body or homogeneous, it becomes important to consider both translational and rotational accelerations. It is likely that the combined effect of both accelerations will be important in predicting the risk of a brain injury occurring.

Existing criteria

To be effective in its purpose, a head injury criterion must provide correlation between the criterion score and risk of injury for a head exposed to the same event conditions. It then follows that for increasing impact severity, the risk of injury would also increase, as would the measured criterion. Conventionally, injury risk is determined by conducting a suite of impact tests using PMHS or animal subjects around the injury threshold severity, or by having detailed information on real world accident cases (where that information is often obtained through reconstructing the accident under controlled conditions and using instrumentation to assess the loading to the test subject). Parameters describing the impact conditions, or criteria derived from those parameters can then used to describe how injury risk changes with changes in impact severity. Statistical analysis will show to what extent the impact severity measure (injury criterion) is able to describe the variation in injury risk.

Unfortunately, the surrogates for living humans used conventionally in injury risk determination are not ideal for the investigation of brain injury. PMHS (Post-Mortem Human Subjects) can be used to study skull fractures and gross mechanical damage to brain tissue and blood vessels; however, there can be no assessment of loss in consciousness and only limited evaluation can be given to any electrochemical disturbance to the axons within the brain. Animal experiments can be used to observe the effects of an impact resulting in minor injuries as the subjects will exhibit concussion and/or temporary brain dysfunction. However, because of complications in scaling results between animals and humans, to account for anything other than size and mass properties (it is not known how to deal with detailed shape differences), such data cannot provide numerical limits directly applicable to humans. Volunteer tests can be used to examine living human responses but these must be limited, ethically, to sub-injurious severities. Finally, real-world accidents can be reconstructed to give information about the impact conditions in relation to the observed outcome. However, in this case the accuracy of the reconstruction is heavily influenced by the information known about the accident itself. Often the knowledge of the accident conditions is far from precise. It is worth noting that accelerometers worn by sports participants may in the future give a very useful and accurate means to study real-world accidents without the need for reconstructions to be used in determining head accelerations and kinematics. It may be possible to obtain acceleration time-histories in three, or even six, degrees of freedom for each impact event.

Peak translational acceleration:

The simplest form of a head injury criterion is to consider purely the peak linear acceleration. The acceleration of a head surrogate is routinely monitored in most forms of test work with specifications in place for sampling conditions and post-event processing. By simply specifying a threshold which cannot be exceeded, engineering solutions to reduce the peak linear acceleration can be encouraged or enforced. Often the peak value is set for the peak resultant acceleration. Alternatively, where coordinate systems are important and defined, it is possible to set direction-dependent thresholds.

Head Injury Criterion (HIC):

In 1972 the US National Highway Traffic Safety Administration (NHTSA) adopted the Head Injury Criterion (HIC). This criterion can be expressed as in Equation 1. In this equation, 't_1' and 't_2' are any two temporal points in the impact (in seconds), and 'a' is the resultant acceleration of the centre of mass of the head (in *g*).

$$\left[\frac{\int_{t_1}^{t_2} a.dt}{t_2-t_1}\right]^{2.5}(t_2-t_1) < 1000 \qquad \textbf{Equation 1}$$

As the HIC has become increasing widely used, it has become subject to extensive evaluation and some criticism.

McElhaney, (2005) identified many factors which are important to head injury but not included in the HIC.

> *"These factors include location of the impact, area of contact,… and angular accelerations induced by off CG [centre of gravity] blows and torso restraint (McElhaney, 2005.")*

As Koch (1988) reported, HIC data is a “mixture of apples and bananas”. This is because a skull fractured in an experiment is likely to give a lower HIC value due to the increased time of deceleration during the deformation and fracture of the skull. Therefore, “HIC values for fractured and non-fractured skulls do not measure the same property.” That is to say, a skull that fractures would not be expected to produce the same HIC value as a skull that didn’t fracture under identical impact conditions. The fracture case will always tend to result in a lower HIC value. This causes increased uncertainty in developing risk functions for head injury where both skull fracture and no fracture cases are included. Around the threshold input level, it may not be the case that risk of injury increases with applied impact severity; if severity is assessed by the HIC or translational acceleration.

When comparing measurements made with a rigid, non-frangible headform, it should be taken into consideration that the headform will tend to give higher translational acceleration and HIC values than would be the case for an identical impact with a human head. Such a difference is likely to be profound for impact conditions where the human head would fracture, thereby attenuating the acceleration.

Prasad and Mertz (1985), as well as Hertz (1993) report the data used to define the HIC skull fracture risk function. The same data are plotted using a Probit regression analysis in Figure 1.

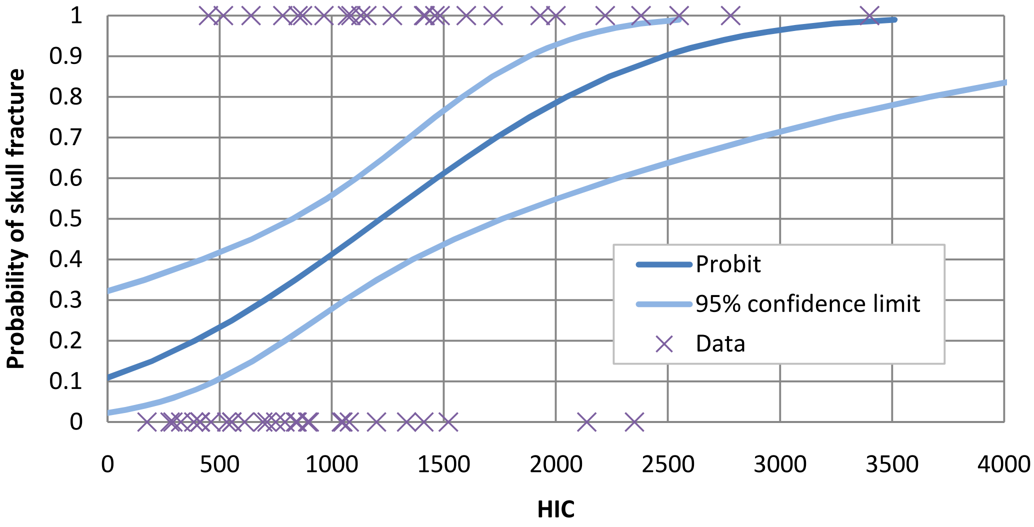

Figure 1: Risk function for skull fracture based on measured HIC_{15} (data from Prasad and Mertz, 1985)

Whilst HIC was principally developed to investigate the risk of skull fracture, Prasad and Mertz (1985) report on the use of HIC in predicting brain injury. A limitation of the injury data is noted in that since only the arterial system was pressurised for the original head impact tests, damage to the venous system was not measured. Therefore venous ruptures, which result in subdural haematomas (AIS 4), would not have been recorded. Neither would it have been possible using these test methods to detect diffuse axonal injuries due to brain cell damage that may result in concussion (AIS 2 to 5); or other brain injuries like cerebral oedema and swelling. On the basis that the absence of an arterial rupture does not signify the absence of brain injury, the resulting HIC risk curve may substantially underestimate the risk of brain injury for a given HIC. A risk curve for the probability of receiving a brain injury based on the HIC data presented by Prasad and Mertz is shown in Figure 2. This was plotted using a Probit regression analysis. Due to the poor significance of the function (Pearson goodness of fit, X^2 = 0.39), confidence limits were not derived. This level of significance, the shallow gradient, and the large intercept at zero HIC show that this function is not reliable for predicting brain injury based on HIC.

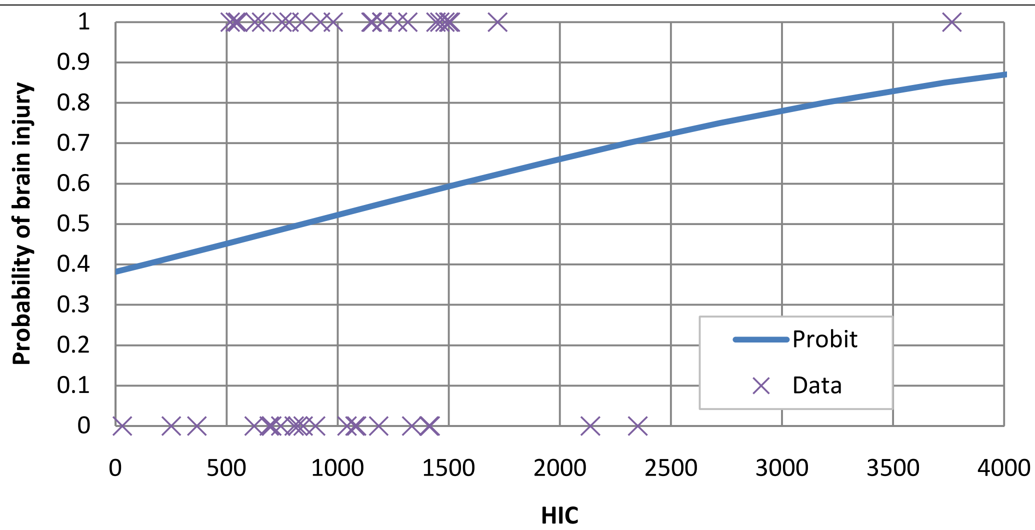

Figure 2: Risk function for brain injury based on measured HIC (data from Prasad and Mertz, 1985)

Prasad and Mertz comment that the risk curves drawn by them using the "Mertz/Weber Method"[1] for skull fracture and brain damage were virtually identical. They infer that for a given level of HIC, skull fracture, brain damage, or both, are equally likely to occur. When their dataset is analysed using an appropriate statistical technique, this inference is not supported. It is worth noting that Gurdjian *et al.* (1949) had already commented that there was no direct correlation between the severity of cerebral damage and linear skull fracture. They reported that a fatality may occur due to brain damage without any skull fracture, and that a skull fracture may occur without any damage to the brain. To be able to consider the risk of skull fracture or brain damage independently, it is therefore necessary to derive specific risk functions.

Maximum rotational acceleration:
Severe whiplash injuries were produced in monkey subjects throughout the 1960s, as reported by Ommaya *et al.* (1967). Ommaya *et al.* proposed a scaling strategy for converting the monkey tolerance to a concussion threshold for man. This scaling was based on discussions with Holbourn, who had proposed a scaling based on brain mass (Holbourn, 1943). One of the key assumptions used in the scaling process was that there is a three dimensional geometric similarity between the brains of a series of subhuman primates and the human brain. However, many other assumptions and caveats are listed by Ommaya *et al.*

The result of the work by Ommaya *et al.* was to suggest that the cerebral concussion tolerance of about 40,000 rad/s^2 observed with rhesus monkeys (note that this tolerance is time dependent) equated to 7,500 rad/s^2 for a human.

Unterharnscheidt (1971) reported on test results investigating tolerance to rotational acceleration. Effects of rotational acceleration on the central nervous system were reported based on experiments with squirrel monkeys. The monkeys were subjected to rotational accelerations of 101 to 386 krad/s^2. A continuum of responses was observed, from no signs of central nervous system damage through concussion to death.

[1] Mertz and Weber (1982) used a modification to the Median Rank method where the input parameter values are ranked from the lowest value associated with specimen failure to the highest value associated with no failure. A line is then drawn between the two end points to provide an estimate of the cumulative distribution of the failure thresholds for the tested specimens.

The potential for this method of generating risk curves to be misleading was demonstrated by Hertz (1993). She showed very similar risk functions for HIC vs. skull fracture for Normal, logNormal and Weibull parametric methods, giving approximately 40% to 45% risk of skull fracture at a HIC of 1000; but when using the Mertz/Weber function for the same data, it gave a risk of <20% at the same HIC level.

Results

Head injuries are not a new phenomenon. For many years they have been occurring frequently in automotive, aviation, sports, and general life accidents. As a result, many investigators have tried to deduce information about the epidemiology of head injury (how head injuries occur) and human tolerance with respect to loading to the head. This means that there is an abundance of published data on head loading cases. To investigate the effectiveness of the existing criteria, a database of head impact cases was constructed based on the information reported in the literature. This database included an assortment of impact types, impact directions, contact surfaces, species (human or monkey), etc. These diverse sources of head impact data were considered to include a broad range of factors that may affect the ability of certain measured criteria, or parameters, to predict head injuries. There were 630 documented cases of head impact testing collated into the database used for analysis.

For 431 cases, the head injury was reported in sufficient detail to determine if it was severe in nature (using an Abbreviated Injury Scale (AIS; AAAM, 2005 {and former revisions}) score of at least 3). For 277 of the cases that were coded as being either AIS < 3 or ≥ 3, a peak resultant acceleration value was also reported (this included cases from boxing, data from the EC sixth framework Aprosys Project, pedestrian accident reconstructions, constrained/forced head motion tests, impactor tests, and full-body drop tests). Based on logistic regression of these 277 cases, the resultant acceleration was shown to be a statistically significant predictor of severe head injury. However, whilst this could predict correctly the absence of a severe head injury in over 70 percent of those events (where most resulted in a non-severe injury or no injury), it could only predict severe head injury in less than half of the severe head injury cases.

A HIC value was reported for 240 of the 431 cases for which an AIS coding was possible (the cases came from the following sources: boxing, the Aprosys Project, pedestrian accident reconstructions, impactor tests, and full-body drop tests). Again logistic regression showed that HIC was a significant predictor of severe head injury. However, this significance was again slightly misleading; as whilst it is true that the HIC could be used to predict the correct result in over 75 percent of the cases, over 80 percent of the severe head injury cases would have been predicted as false negatives.

An obvious issue that was noticed with the conglomeration of head injury cases was that the monkey subjects had been exposed to much higher acceleration levels than the human subjects. Therefore the logistic regression analyses were repeated looking at human subject cases only.

For the human subjects (data from PMHS tests as well as accident / accident reconstruction data) the peak resultant linear acceleration was again significant in predicting the severity of head injury. Again, whilst it correctly predicted an AIS < 3 result in almost 95 percent of those cases, the AIS ≥ 3 result was correct for only about one third of the severe head injury cases. The HIC gave only a very slightly improved predictive ability over the raw resultant linear acceleration. Despite predicting more of the AIS < 3 cases correctly, the peak resultant angular acceleration was the measure least able to predict the AIS ≥ 3 injury outcome correctly.

A graphical comparison of the predictive capabilities of the peak resultant linear acceleration, resultant angular acceleration, and HIC is shown in Figure 3, for the human head impact cases.

Of the human cases, 162 had both an AIS coded head injury and a peak resultant angular acceleration value (the cases came from the following sources: Aprosys Project, boxing, pedestrian accident reconstructions, impactor tests, and full-body drop tests). For the 125 cases of AIS < 3 head insult the rotational acceleration value could be used to predict that result correctly for all but three of these (almost 98 percent). However, in only two of the 37 AIS ≥ 3 cases would the rotational acceleration have given the correct result, instead suggesting AIS < 3.

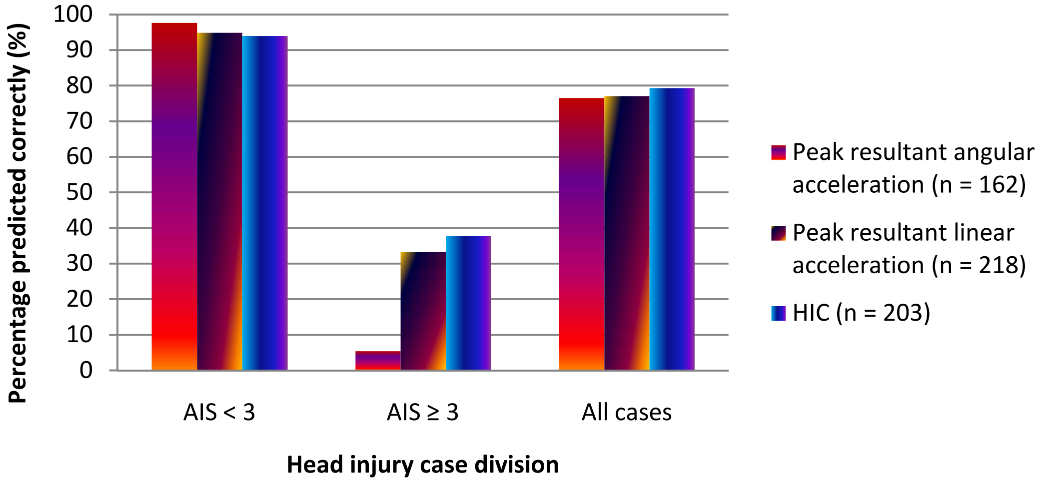

Figure 3: Percentage of human head impact cases where the outcome was predicted correctly by the linear or angular accelerations, or HIC

One criticism raised regarding the use of the HIC is that skull fracture can affect heavily the measured peak linear acceleration and HIC value. A fracture will allow greater deformation of the skull than would occur without a skull fracture, thereby attenuating the acceleration under equivalent impact conditions. To investigate this effect, the presence or lack of a skull fracture was included as a covariate in the logistic regression of HIC with AIS ≥ 3 head injury (the sources of data for these cases were: boxing, Aprosys Project, pedestrian accident reconstructions, impactor tests, and full-body drop tests). The results suggested a significant influence of skull fracture, and an improvement in the predictive capability accounting for skull fractures. However, for the no skull fracture cases, the HIC still only predicted one of the nine AIS ≥ 3 cases correctly.

The influence of a skull fracture on the ability to predict AIS ≥ 3 injuries suggests that one must know the skull fracture outcome for a case before considering the brain injury outcome. This will obviously cause issues when applying HIC to prospective cases (e.g. research testing with a crash test dummy); although, this would not be expected to cause a problem where the head loading was not going to cause a skull fracture, for example in non-contact cases or at sub-fracture loading levels.

The likelihood of a skull fracture occurring for a given HIC value was investigated using a Cox regression survival analysis, as shown in Figure 4 (human head impact data were taken from boxing impacts, and full-body drop tests). As is the case for each of the four curves shown (flat or shaped impact surface and padded or rigid impacts), the probability of a head surviving without a fracture decreases with increasing HIC. Each experimental outcome influences the risk of skull fracture, which is the reason for the many changes in gradient, exhibited by each of the lines.

For the purpose of this investigation, the helmeted conditions used by Got *et al.* were considered as padded (Got *et al.*, 1978, and Got *et al.*, 1983). From Figure 4 it can be seen that there is a protective effect of padding, and impacts with a flat rather than curved surface. This is likely to be due to those two conditions spreading the impact force over a greater area of the skull; assuming that impact pressure or stress is related to the likelihood of fracture. Padding will also help to attenuate impact force, to some extent, through energy absorption.

Impact site (i.e. frontal, side, rear, etc.) did not significantly affect the ability of the HIC, or peak resultant accelerations to predict AIS ≥ 3 head injuries. However, impact site was significant when looking at subsets of the data. For instance, the data of Ono *et al.* (1980), based on tests with monkey subjects, showed that frontal or rear impact directions were a significant ($p = 0.02$) factor when using the peak resultant linear acceleration to predict AIS ≥ 3 head injury.

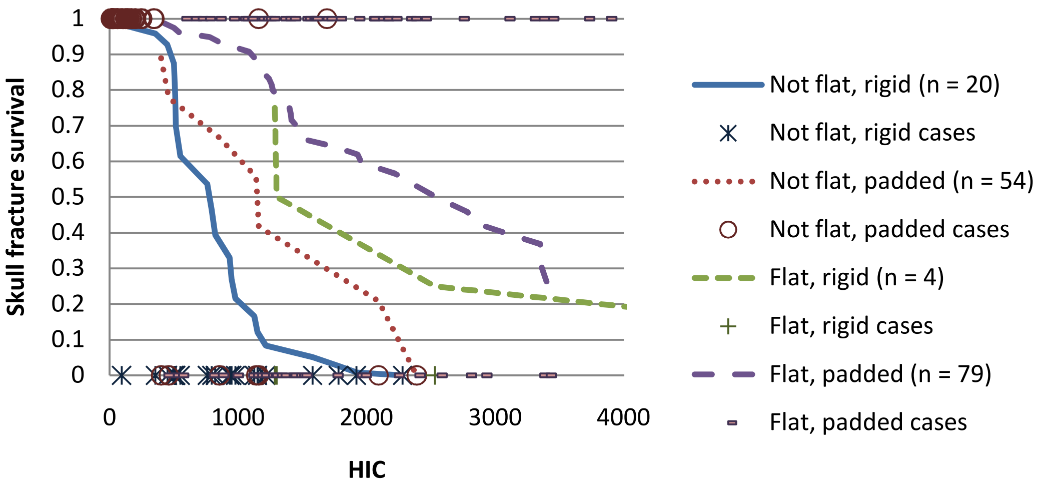

Figure 4: Cox regression plot for probability of avoiding a skull fracture with HIC, for human head impact cases

Advanced head injury criteria, which incorporate both linear and rotational head accelerations, have already been proposed. The GAMBIT and the HIP were reviewed as part of this study. However, neither of these criteria have been sufficiently validated or evaluated to make them appropriate for widespread use at this time. It has not been possible to provide a detailed assessment of them within this investigation (GAMBIT and HIP values were not reported in the historical literature and cannot be calculated reliably from the peak head acceleration values that are generally documented). Instead it seems as though data from throughout the full impact time history, in the six degrees of freedom, three orthogonal linear and rotational axes) will be necessary to use these criteria effectively. New sources of information that contain such complete data together with detailed descriptions (including extent, location, and severity) of any injuries are not readily forthcoming at this time. Such data is needed in support of advanced head injury criteria.

Discussion

From the historic data it has not been possible to prioritise factors contributing to the head injury risk because not all factors are controlled for, or reported in, all of the studies. This means that when considering all contributing factors together, little, if any, statistical power has been gained through combinations of the data in the literature.

Throughout the analysis of data from the literature, there has been no clear advantage shown when using either peak linear resultant acceleration, HIC, or peak rotational resultant acceleration to predict head injury outcome at the AIS ≥ 2 or AIS ≥ 3 levels. The following summary figures (Figures 5 and 6) show the prediction efficacy of the linear or rotational acceleration, or HIC for either AIS ≥ 2 or AIS ≥ 3 head injuries to humans or squirrel monkeys.

One of the main limitations with the assessment of injury outcome predictions is the data used for the evaluation. The injury information is only what can be obtained from the reference sources reviewed. Typically, within the historical data the focus has been on the more severe of the spectrum of head injuries. This is particularly true when considering the PMHS impact cases where it is difficult to identify clearly anything but severe brain injuries and skull fractures. Hence from Figure 6, the human AIS ≥ 3 prediction outcomes are closer to those from the squirrel monkey tests than in Figure 5 for AIS ≥ 2 head injuries. It should be remembered, at this stage, that relatively simple closed fractures of the skull are AIS 2 injuries. The human dataset used to derive Figure 5 will include many cases of skull fracture (an impact loading injury), whereas the squirrel monkey injuries at the AIS 2 level are more likely to be intracranial brain injuries, from inertial loading of the head. Rotational acceleration is not expected to be a good predictor of skull fracture. This may explain part of the reason why the prediction of human AIS ≥ 2 head injury outcome, based on rotational acceleration, is not as good as the predictive ability shown with the squirrel monkey data.

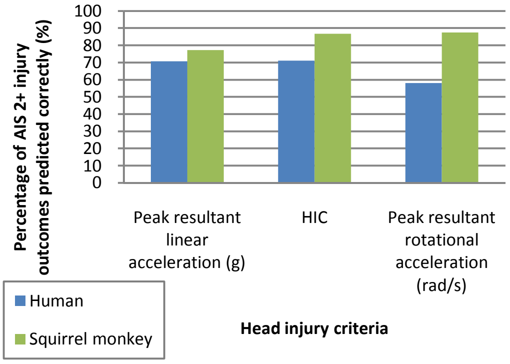

Figure 5: Head injury outcome prediction efficacy for AIS ≥ 2 injuries

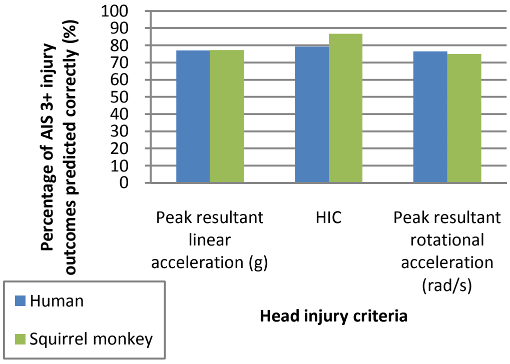

Figure 6: Head injury outcome prediction efficacy for AIS ≥ 3 injuries

Based on logistic regression of a subset of cases from the full dataset, the resultant linear acceleration was shown to be a statistically significant predictor of severe head injury. Peak linear head acceleration and the Head Injury Criterion (HIC) were found to be reasonable predictors of serious (Abbreviated Injury Scale, AIS ≥ 3) head injury occurrence. However several features of the impact conditions were shown to contribute to the injury outcome. Therefore, whilst these criteria seemed useful if applied in a very general way, specific thresholds would have to take account of the confounding factors such as the impact conditions. This supports the real-world observations that using linear head acceleration based criteria to control the loading to the head in a general way is a useful first step in the mitigation of head injuries.

It is clear, considering all sorts of biomechanical data, that limiting the force, concentration of force, and the linear and rotational accelerations to which a head is subjected is a sensible approach when trying to reduce injury risk. These basic engineering principles have guided the development of: test tools to assess the severity of head impacts, test procedures, and basic injury criteria for the use in testing potential road-safety related head impacts. Such testing has facilitated the development of injury prevention countermeasures such as padded vehicle interiors, airbags, cycle and motorcycle helmets, etc. As a result, since the implementation of testing, through regulatory or consumer testing schemes, they have driven improvements in head protection levels. These head injury countermeasures have been effective. For instance, the incidence of severe head injuries in frontal impact car crashes has reduced substantially since the widespread introduction of airbags (e.g. Knack *et al.*, 2004).

The proposals to use peak linear head acceleration or HIC to protect against head injuries were based on testing relevant to a specific historic need. Moving on from that situation, it is likely that head injury prevention will need to address diffuse injuries occurring away from the point of impact. Based on their derivation and limits of application, peak linear acceleration and HIC do not seem sufficient for that purpose. This may be particularly true for moderate severity injuries or in predicting precisely the type of brain injury. Therefore, additional risk assessment tools are likely to be necessary.

The rotational acceleration components of a head impact are likely to have a greater bearing on intracranial head injury than skull fracture (e.g. Gennarelli *et al.*, 1972). Also, an inference from the APROSYS Project work (Deck *et al.*, 2007) could be that it is important to consider the combined loading to the head (rotational and linear accelerations) when estimating what type of intracranial injury is likely to occur. This is a reasonable assumption, that simple consideration of the linear acceleration of the head (peak resultant value, or HIC) will not provide an accurate prediction of intracranial head injury occurrence. Consideration of the linear head kinematics alone is unlikely to be sufficient to judge the most likely type and location of potential intracranial head injuries. This is particularly important when considering those injuries with a low threat-to-life, such as concussion and mild diffuse axonal injuries. Reasoning such as this has fuelled, and continues to support, the development of advanced finite element models of the head. However, having an injury criterion based on global head motion and relating to intracranial injuries should still offer a substantial benefit. It would bridge the gap from routine use of the HIC and peak resultant linear acceleration to the time when advanced head models can be used as regular tools for head injury risk testing.

In order to be able to develop advanced head injury criteria, or validate a detailed head model irrefutably, for assessing brain injury risks, it is necessary to have head injury data sources that consider both, not either, rotational and translational head kinematics. Peak acceleration values alone are not sufficient as the timing, duration, and strain rate of the forces to which the head is subjected are also likely to be important. Having full acceleration time-histories is likely to be particularly important given the assumption that combinations of rotational and translational accelerations are necessary for improved injury criteria. The combinations should consider criterion values at each point in time, not just the isolated peak values from any time throughout the acceleration pulse.

As biomechanical research continues to investigate the detailed aspects of head injuries it becomes apparent that the traditional criteria, or indeed any new criterion, would need additional validation (beyond that available from the published data). This validation would have to account for the conditions of loading to the head and the particular head injury under consideration.

Conclusions

Published head injury research from the past 70 years has been reviewed to:

- Collate existing propositions with respect to mechanisms for head injuries
- Draw together a dataset of head injury case data based on published information
- Compare this dataset with expected results based on previously published assertions (making use of statistical techniques for this analysis)

Even with a broad dataset of over 600 head injury cases collated from the literature, it has not been possible to prioritise, according to their importance, the impact factors that contribute to head injury risk. Neither has it been possible to develop an improved estimate of head injury likelihood, which takes into account these contributory factors. One reason for this is that not all factors were documented for all studies conducted previously. Due to the extent of the missing data only limited advantages have been observed through combining the historic data.

Peak head acceleration and the HIC are reasonable predictors of serious (AIS $\geq$ 3) head injury occurrence, based on historical test data. However several features of the impact conditions have been shown to contribute to the relationship between the injury predictor and the injury outcome. These are:

- Occurrence of a skull fracture
- Impact surface padding
    - Padding affected risk of skull fracture
- Profile of the object impacted
- Impact site
    - Demonstrable with a small subset of animal test data only

To address other diffuse injuries occurring away from the point of impact, additional risk assessment tools are likely to be necessary.

The analysis of the published head injury case data was not able to show the benefit of including a rotational acceleration component in the injury risk assessment. This was due to a number of reasons:

- Unreliable brain injury detection
- Inadequate reporting to determine the occurrence of less life threatening injuries
- Linear and rotational acceleration data not always reported
- When reported, only peak data values given in the literature

To show the usefulness of injury criteria that account for rotations as well as linear components additional data are needed. Equally, to derive an advanced head injury criterion, or validate irrefutably a detailed head model, for assessing brain injury risks, new data would be required. These data would need to include:

- A detailed description of the injuries occurring
    - Including both the type and location of the injury
- Rotational and linear acceleration data, throughout the time series of the impact event

Until such data are available, suggestions for new criteria will be subject to the limitations of the existing data. These limitations have been shown to be manifold.

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