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The potential and limitations of utilising head impact injury models to assess the likelihood of significant head injury in infants after a fall

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Abstract

The use of engineering principles in assessing head injury scenarios is of increasing significance in investigations into suspected child abuse. A fall scenario is often given as the history for a head injury to an infant. This paper addresses the basic engineering principles and factors to be considered when calculating the severity of a head impact after free-fall. The application of head injury models (HIMs) to ascertain the forces involved in childhood head injuries from impact is also discussed. Previous studies including Duhaime et al. [J. Neurosurg. 66 (1987) 409] and Nokes et al. [Forensic Sci. Int. 79 (1995) 85] have utilised HIMs for this purpose: this paper reviews those models most widely documented.

The HIM currently considered the 'state-of-the-art' is the head injury criterion (HIC) and it is suggested that this model should be utilised for assessing head impact injury in child abuse cases where appropriate.

Keywords: Head injury; Biomechanics; Head injury models; Child abuse; Childhood falls

1. Introduction

Establishing whether a head injury to an infant was the result of accident or abuse is a fundamental problem in forensic investigation. A fall scenario is often given as the cause of a head injury. Until recently the legal system relied on the testimony of medical experts to determine whether the force imparted to the head in a given scenario (e.g. a fall from a specified height) was consistent with the resulting head injury, even though there might be little scientific basis for their conclusions.

Since the 1960s engineers have attempted to develop mathematical models for the mechanisms of head injury. The models utilise fundamental Newtonian principles and build upon experimental observation to develop prediction equations to estimate the likelihood of significant injuries for a given scenario. This paper provides an overview of basic fall mechanics and examines the more popular prediction models for head injuries developed over the last three decades.

2. Investigating the severity of a fall

Many factors have to be considered when investigating infant injuries possibly sustained from falls. These parameters will affect the severity and type of injuries sustained during a fall and include acceleration due to gravity, air resistance, height of the fall, impact velocity, fall and impact mechanics, impact surface, age and physical condition of the infant.

2.1. Acceleration of a free-fall due to gravity (g)

The acceleration (a) of a free-fall is constant due to gravity ($g = 9.81$ m/s^2) and linear, that is in a straight line from the initial fall position towards the point of impact (Fig. 1). Acceleration is defined as the rate of change of velocity with respect to time (distance/time^2).

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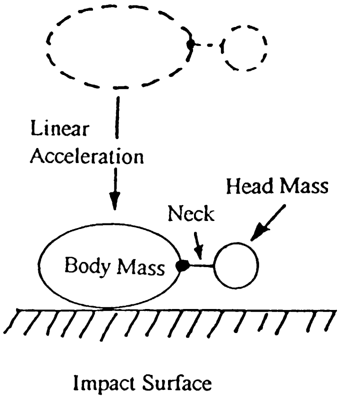

Fig. 1. Linear acceleration.

2.2. Air resistance

A body free falling under the effect of gravity in a vacuum will continue to accelerate unopposed. However, a body falling under the effect of gravity in air will experience an opposing force provided by the resistance of air. A more streamlined fall, i.e. feet or head first, would cut through the air more efficiently thus reducing the effects of air resistance. Conversely, a fall in the prone or supine position would increase air resistance. These differences relate to the surface area of a body that is moving through the air.

Air resistance opposes the gravitational acceleration, which in turn reduces the velocity (m/s). Snyder [3] presents graphs showing velocity at impact plotted against free-fall height calculated with and without the effects of air resistance. Air resistance is shown to have an insignificant effect on impact velocity during free-falls from less than 50 ft (15.24 m).

2.3. Velocity on impact (v) and fall height (s)

The velocity is the speed of a body in a given direction. A positive or negative sign denotes the direction of a fall. For example, where the velocity of a falling body increases, accelerating at 9.81 m/s^2, at impact the direction of the velocity is denoted as $+V$ (m/s). The positive sign defines the direction as downward. At some point during the impact the velocity of the body would decrease to 0. If the body rebounded up from the surface then the velocity would be $-V$ (m/s) before again decreasing to 0 as it came to rest. The negative sign defines the direction as upward. Velocity directions should always be defined in calculations to avoid incorrect assumptions. Often the direction is not defined by a positive or negative sign. The direction may be implied from words, for example, if the velocity of a falling body is given, it can be assumed that the direction is towards the impact surface.

Assuming linear acceleration and ignoring air resistance the velocity at impact is directly related to the fall height by Newton's equation of motion, i.e.

$$v^2 = u^2 + 2as$$

where v is the final velocity (m/s), u the initial velocity (m/s), a the acceleration due to gravity, $g = 9.81$ m/s^2, and s is the distance (fall height) (m).

In most falls the initial velocity (u) is 0, i.e. before the fall occurs the body is not travelling in a vertical direction, so the formula can be simplified:

$$v^2 = 2as$$

The higher the fall the greater the velocity at impact (Example 1).

Example 1. Calculating the velocity (v) for different fall heights (s).

A fall from a height of 10 m would give a velocity at impact of

$$v^2 = 2as$$

$$v = ?,\ a = g = 9.81\ \text{m/s}^2,\ s = 10\ \text{m}$$

$$v = \sqrt{2 \times 9.81 \times 10},\ \text{or}\ v = 14.0\ \text{m/s}$$

Using the above formula, a fall from a height of 1 m would give a velocity at impact of 4.4 m/s.

2.4. Mechanics of a fall

Falls involving infants may occur in a number of ways including a baby rolling off a changing mat on a table, rolling off a bed or falling from a carer's arms. Falls of this nature involve only vertical linear acceleration since the horizontal component may be assumed to be negligible. The velocity on impact can be calculated using the formula $v^2 = 2as$ (Fig. 2, Example 1).

Some falls may involve an initial increase in height from an elevated surface, for example, a jump before a fall or an infant thrown into the air. Each would increase the height of the fall and subsequently the velocity on impact (Fig. 3, Example 2).

Example 2. The relationship between the impact velocity (v) and a fall involving an initial jump or throw (s).

From Fig. 3, where $s_1 = 0.2$ m and $s_2 = 1$ m:

$$s = s_1 + s_2 = 0.2 + 1 = 1.2\ \text{m}$$

therefore:

$$v^2 = 2as,\ \text{or}\ v = \sqrt{2 \times 9.81 \times 1.2} = 4.8\ \text{m/s}$$

The velocity at impact increases as the height increases.

If a person was running just before the fall occurred, for example, along an elevated surface such as a wall, and slipped off, the initial horizontal velocity will not be 0 and the horizontal displacement will be increased. If the

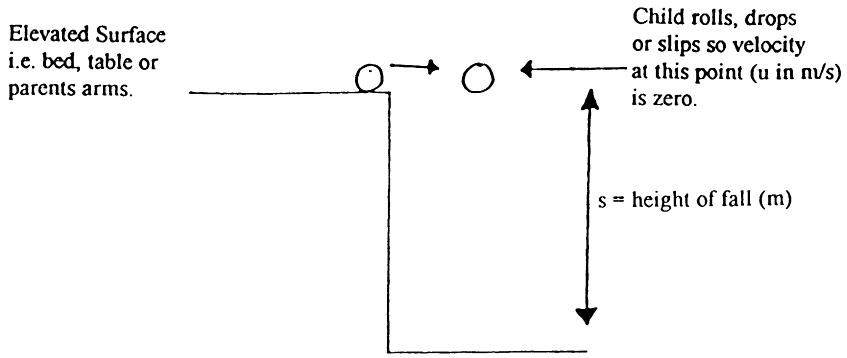

Fig. 2. A fall involving linear acceleration.

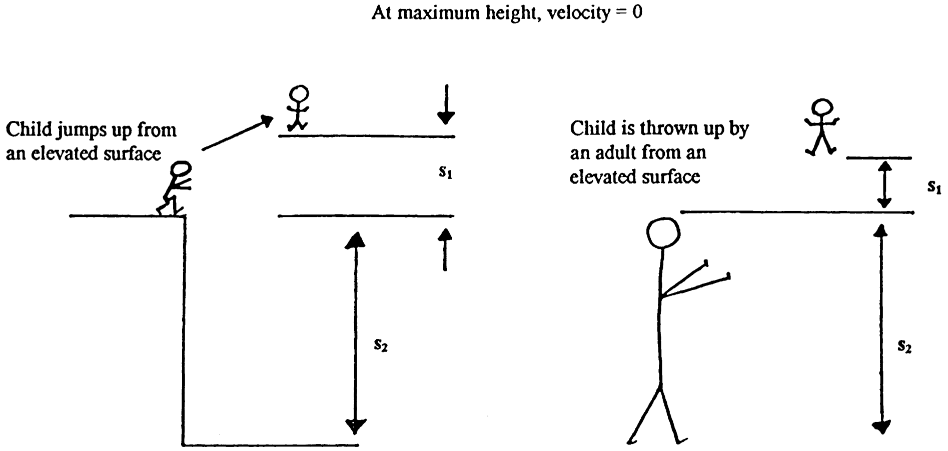

Fig. 3. A fall involving an initial jump or throw.

horizontal velocity of the body at impact is greater than 0 a sliding or skidding type of impact may occur (Fig. 4, impact A). However, horizontal velocity will decrease and may reach zero during the fall (Fig. 4, impact B). Apart from the increase in horizontal displacement this type of impact (B) would be very similar to that illustrated in Fig. 2.

2.5. Deformation characteristics of the impact surface and the mechanics of impact

The impact surface influences many factors such as absorption of kinetic energy, acceleration on impact, impact duration, impact force and area of contact. These last two factors determine stress (force/unit area).

2.5.1. Kinetic energy (KE)

A body acquires KE[1] as it is falling. Almost all of the energy is transferred to and absorbed by the impact surface and/or the impacting body. The degree to which an impact surface deforms affects how much KE is absorbed by the surface and thus the amount that remains to be absorbed by the body.

The relationship between KE, velocity (v), force (F) and mass (m) of the impacting body and the stopping distance (or deformation distance, d or s) is illustrated in Fig. 5.

Newman [5] suggested that the simplest relationship between the forces produced and the stopping distance required to absorb energy is

$$Fd = \tfrac{1}{2}mv^2 = \mathrm{KE}$$

where d is the stopping distance (m), F the average force during the impact (N), v the velocity (m/s), m the mass of impacting body (kg), and KE is the kinetic energy (J).

From the above relationship:

$$Fd = \mathrm{KE}, \quad \text{so}\, F = \frac{\mathrm{KE}}{d}$$

i.e. a greater stopping/deformation distance reduces the force generated on impact and reduced force causes less damage to the impacting body.

[1] Kinetic energy (KE) — measured in joules (J). The energy of motion, usually defined as the work done by a body possessing energy when it is brought to rest [4].

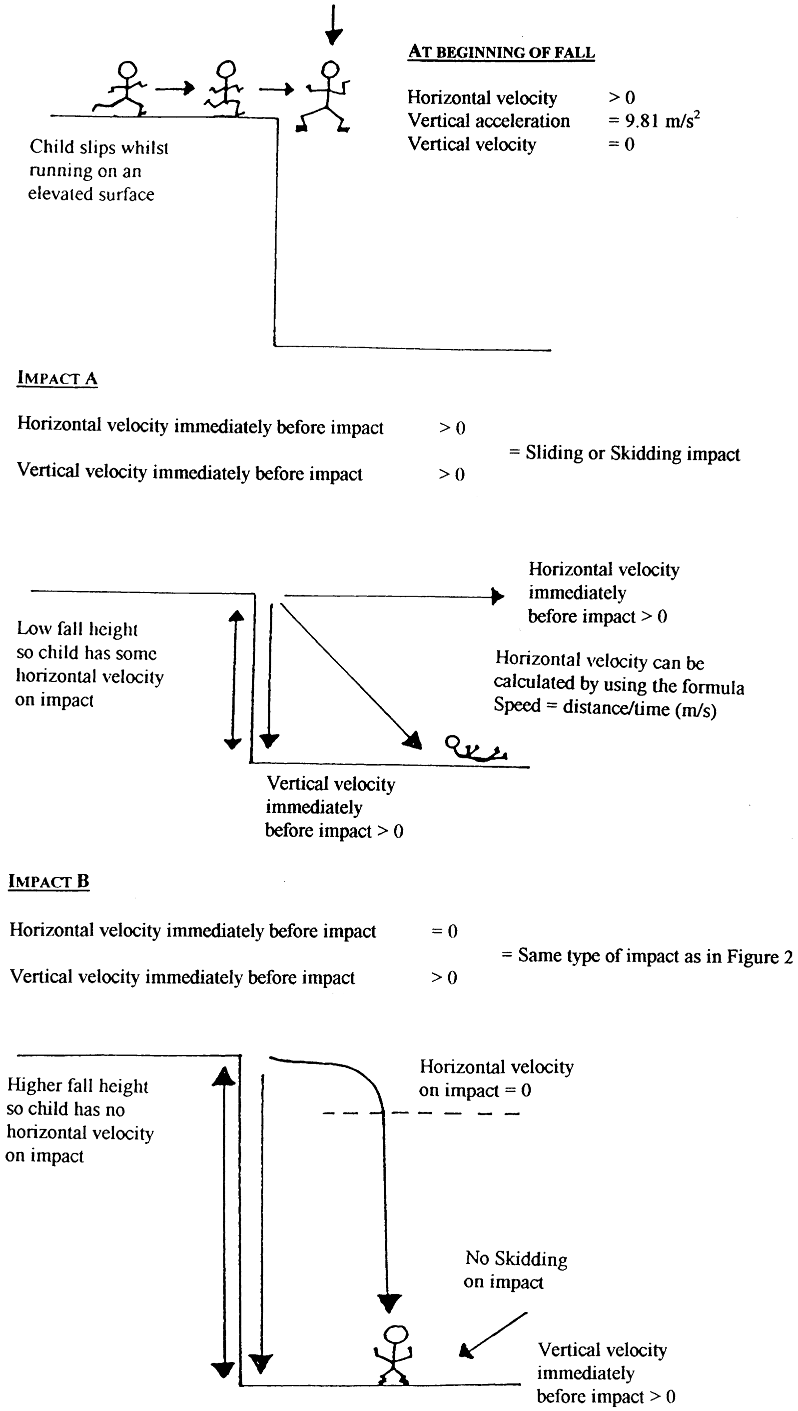

Fig. 4. A fall involving a previous forward (horizontal) motion.

2.5.2. Acceleration

Acceleration is the rate of change of velocity and like velocity it is directional, denoted by a positive or negative sign. It is usually expressed in g, where $g = 9.81$ m/s^2. A body falling under gravity will continue to accelerate unless acted on by an external force. If the body is subjected to an external opposing force, for example, if it strikes a surface perpendicularly, acceleration will be negative. Alternatively,

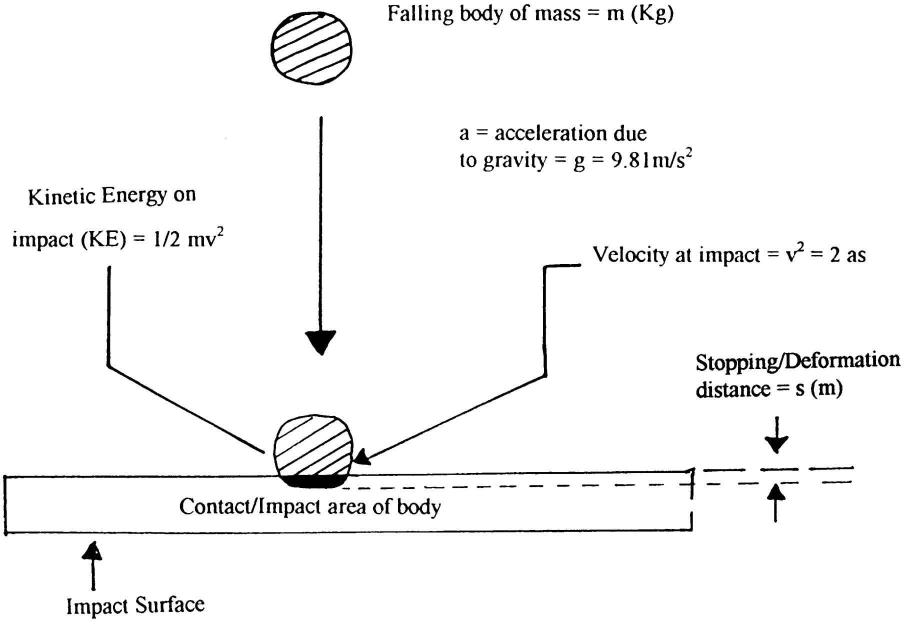

Fig. 5. The relationship between kinetic energy, velocity, force and mass of an impacting body and the stopping, or deformation distance (s).

the body can be said to decelerate. The terms acceleration, deceleration and negative acceleration are used interchangeably. The acceleration of the human body during impact will influence the severity of injuries sustained. To reduce injury it is necessary to reduce the magnitude of the peak acceleration — the greatest rate of slowing down — which occurs on impact and this can be achieved by allowing the body a greater distance and time over which to stop. A surface that cushions an impact will provide a longer impact duration than a rigid surface. If the surface is rigid the deformation distance is small so that the body is brought to rest over a short time period. This means that the peak acceleration will be increased giving the possibility of more severe injury. Therefore, during a head-first fall the severity of head injury is highly dependent on whether the impact surface is rigid (e.g. concrete) or deformable (e.g. rubber) (Fig. 6).

The acceleration of a body on impact is related to the impact force by the equation

$$F = ma$$

where m is the mass of impacting body (kg) and a the acceleration (m/s^2).

This equation can also illustrate how a greater deformation distance reduces impact force (Example 3). King and Ball [6] showed how "in terms of g, the acceleration of the falling body on impact can be expressed as h/s, the ratio of the fall height to the deformation distance".

$$a = \left(\frac{h}{s}\right)g$$

where a is the acceleration (m/s^2), h the fall height (m), and s the deformation distance (m).

Example 3. The relationship between stopping distance (s), acceleration on impact (a) and the force on impact (F).

If an 80 kg body is dropped through a distance of 10 m onto two surfaces with stopping distances 0.02 and 0.01 m, respectively:

$m = 80$ kg, $h = 10$ m, $g = 9.81$ m/s^2.

Using $a = (h/s)g$:

If

$$\text{(i)}\ s = 0.02\,\text{m} \Rightarrow a = \tfrac{10}{0.02}g = 500g$$

$$\text{(ii)}\ s = 0.01\,\text{m} \Rightarrow a = \tfrac{10}{0.01}g = 1000g$$

Then

$$F = ma$$

$$\text{(i)}\ F = 80 \times 500 \times 9.81 = 392.4\,\text{kN}$$

$$\text{(ii)}\ F = 80 \times 1000 \times 9.81 = 784.8\,\text{kN}$$

the surface with the greater stopping distance $s = 0.02$ m produced a lower impact force = 392.4 kN.

2.5.3. Stress

If, during an impact event, a surface deforms (alters in shape) energy is absorbed by the work done to deform the material. If a surface is curved or irregular, deformation may increase the impact/contact area between the body and the surface and dissipate the force over a larger area, therefore

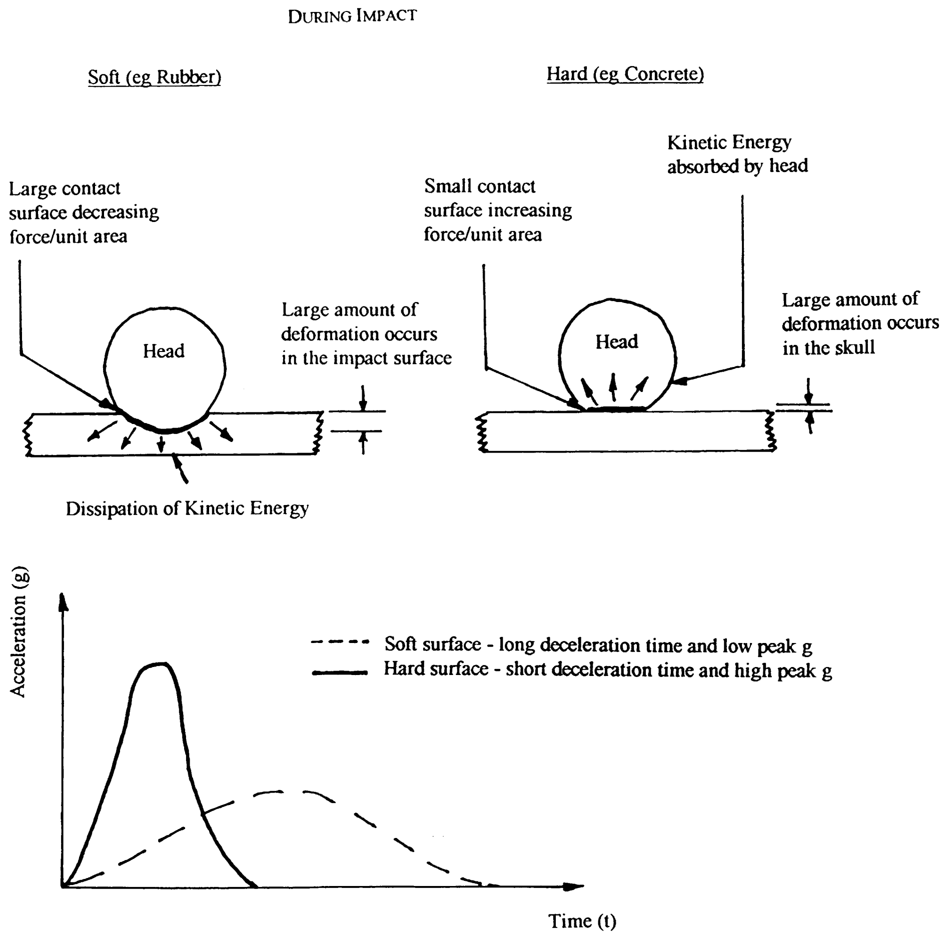

Fig. 6. Impact of the skull onto deformable (soft) and rigid (hard) surfaces.

reducing the force per unit area (stress) on the body and thus, the severity of injury (Fig. 6).

2.5.4. Impact duration

Snyder [3] suggested that an increase in impact duration reduced injury. Actual fall cases were listed to illustrate the suggestion concluding that "the resilient qualities of these impacted materials, which deformed over a relatively long time period, were evidently quite effective in decelerating the body with minimal, or no injury, in these cases".

$$t = \sqrt{\frac{2s}{a}}$$

where t is the impact duration (seconds), s the deformation distance (m), and a the acceleration (m/s^2).

2.6. Orientation of the body on impact

The amount of energy absorbed by the impact surface and other areas of the body dictate the amount of energy absorbed by the head. A feet-first impact with secondary impacts to the shoulder and finally the head may result in the feet, legs and shoulder absorbing a greater proportion of the impact energy. Similarly, in a head-first fall where the head is the first part of the body to strike the surface, the head would absorb a greater proportion of the impact energy. Therefore, this type of head first fall would be more likely to cause a life threatening brain injury than a feet first fall.

Nimityongskul and Anderson [7] studied 76 falls involving children under 16 years of age (75% were 0–5 years old) and found that "most injuries occurred to the head and face region. In climbing out of bed or crib, the child is likely to fall 'head first', perhaps because there is greater weight in the upper half of the body, especially the head". Smith et al. [8] study (age range 4 months to 15 years) also suggested that the relatively large head in children "tends to make the centre of gravity higher than in adults so that they pitch forward in a fall". Collantes [9] also commented that "the lack of co-ordination and the disproportionately large head mass add to the child's tendency to fall in a head-first

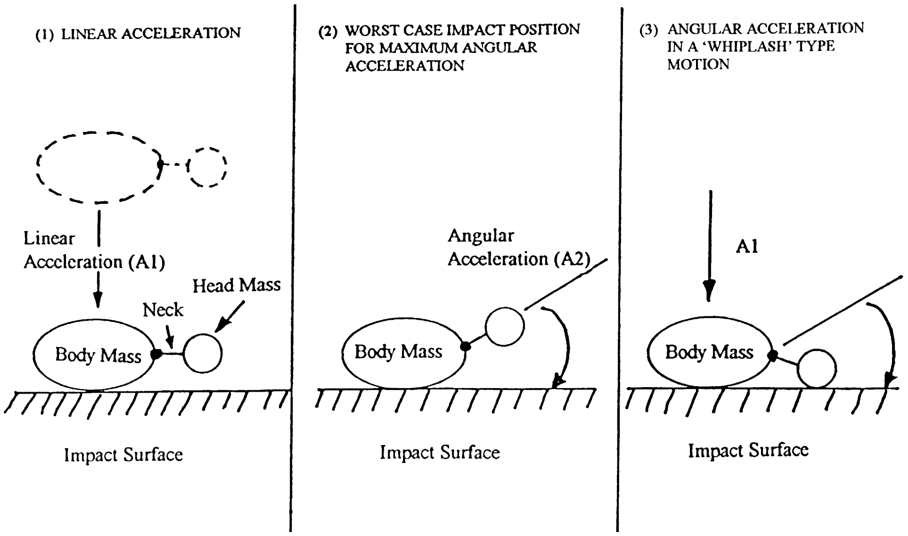

Fig. 7. Angular acceleration of the head due to whiplash.

orientation". In a pure free-fall the larger head mass would not alter the body orientation, as acceleration is independent of mass.

During a fall the body is accelerating due to gravity (linear acceleration A1, Fig. 7(1)). However, other accelerations can act on the head during the impact event. If the primary impact involves the bulk of the body with the neck in a hyper-extended position (angular acceleration A2, Fig. 7(2)), the secondary impact will be that of the head in a whiplash type of motion. Thus, angular acceleration of the head is produced.

Other authors may refer to angular acceleration of the head as rotational acceleration.

This mechanism could occur in any portion of the body, e.g. above the waist where the initial impact is with the buttocks or the whole body with a feet first impact. The portion of body above the first point of contact could be subjected to an angular acceleration where the contact point acts as the centre of rotation about which the rest of the body accelerates in an arc. The net result is that the body is subjected to both linear and angular acceleration.

Also axial rotation (rotational acceleration A3, Fig. 8) of the head could be caused during the same impact scenario. Axial rotation is defined (using the standard anatomical position) as the rotation of the head in the horizontal plane. As the head is accelerating towards the impact surface it may also be subjected to some axial rotational acceleration.

2.7. Age

During the first years of life infants are believed to be more prone to serious head injuries from mechanical abuse, their skull and soft membranous fontanelles providing little or no protection to the brain from external forces [10].

It has been suggested that "... the unfused sutures and softness of the calvarium" (*sic*) "allow for more brain movement ..." [11] and "the thin, partly membranous calvarium" (*sic*) "is supple and permits easy stretching of the brain and its veins" [12]; such factors may make the infant brain more susceptible to injury from mechanisms such as falls, and impacts from blows and shaking.

Subdural haemorrhage "can occur at any age, but is common at both extremes of life" [13]. It has been said that the fluid filled spaces between the meninges of the brain are larger in infants than adults. Aoki and Masuzawa [14] cite a paper by Sanada et al. [15] suggesting that "in 7–8-month-old infants, it has been demonstrated that the subarachnoid spaces are physiologically widened". Brown and Minns [16], citing Nickel and Gallenstein [17] suggested "... if the subarachnoid space is greater than normal (as in old age, or in the first year of life when it is up to 1 cm deep), more swirling [rotation] is possible. These are pre-disposing factors to subdural bleeding". These suggestions are very important when considering brain injury in infants as the presence of a

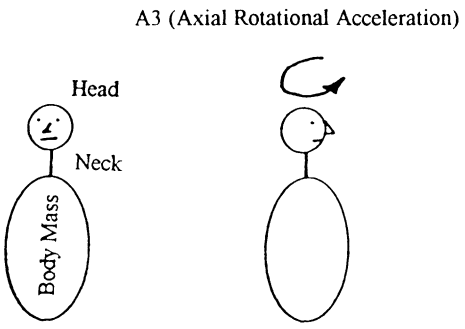

Fig. 8. Axial rotational acceleration of the head.

larger fluid filled space could increase the distance between the brain and skull which may, during mechanical trauma, increase the likelihood of injury to 'bridging veins'.

At the first point of contact the skull will strike the surface and begin to decelerate rapidly due to its rigid structure. The brain also begins to decelerate but at a much slower rate due to the soft pliable nature of the brain and the cushioning of the cerebrospinal fluid. The disparity between the initial rates of acceleration experienced by the skull and brain may be referred to as "deceleration lag": this lag may cause stretching and rupture of the bridging veins that connect the brain to the skull which leads to rupture and bleeding into the subdural 'space' (Acute Subdural Haematoma) (Fig. 9).

2.8. Measures of injury severity on impact

There are many variables which may cause injury during impact: the kinetic energy of the body, the force imparted to the body, the stress and strain of the bones and tissues, and the overall acceleration of the body. For every fall a different proportion of each of these factors is responsible for the injuries.

Gadd, in 1966, suggested [18] that although the body is exposed to these other variables under impact, "it has been impractical except in very limited instances to obtain transducer readings ... which are directly associated with the injury, and, as a result, the overall head acceleration, a rather indirect measure, has come into wide use" [18]. This measure can be utilised to test products such as safety helmets, the inner structure of cars, playground surfacing and other products with the aim of reducing the likelihood of a major head injury.

Acceleration can be calculated for a particular fall/impact scenario using the formulas given in Section 2. Although Newtonian analysis [2] of head impact can be useful in approximating the severity of head injury from a fall, it has

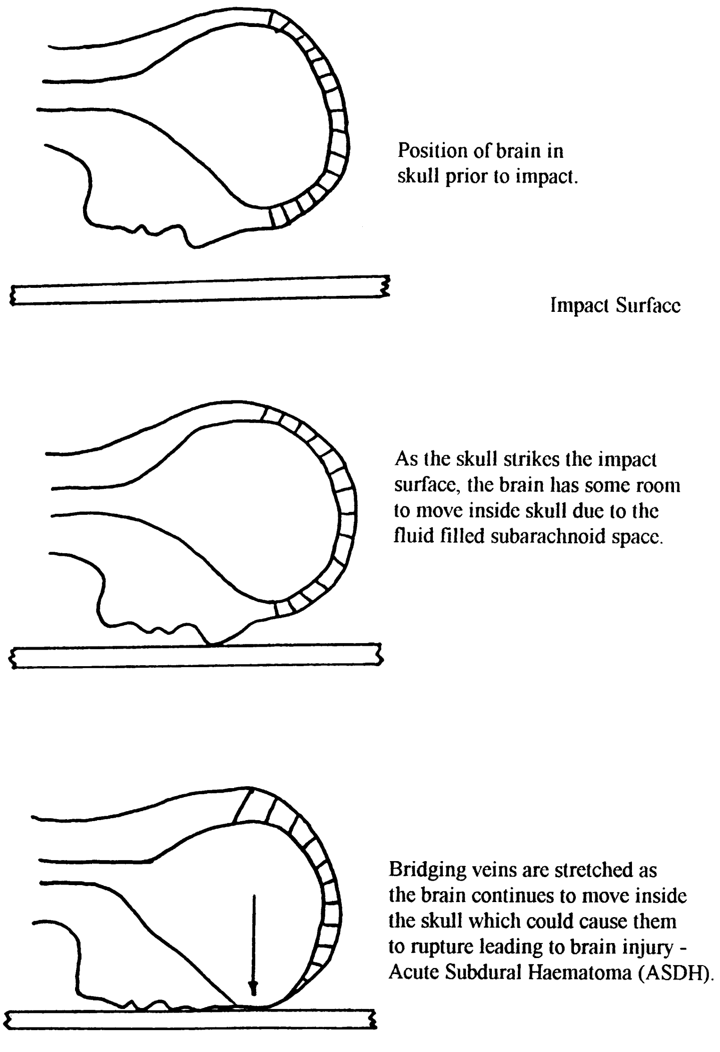

Fig. 9. The deceleration lag of the brain after the skull impacts a surface.

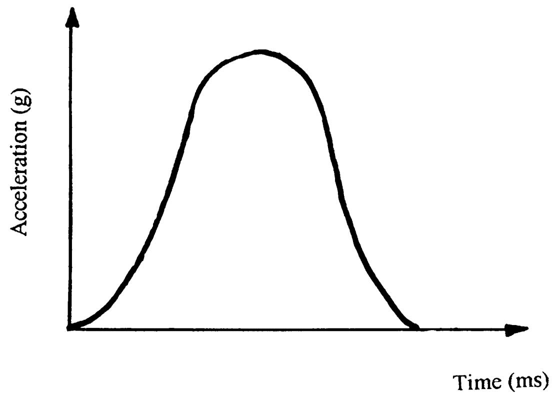

Fig. 10. A typical *g*/time trace of an impact.

been shown [19] that this analysis neglects to ascertain the impact attenuation properties of the impact surface especially where the surface comprises more than one material, e.g. linoleum laid on floorboards or carpet laid on concrete. Therefore, it has been recommended [19] that where possible an in-situ simulation should be conducted in order to assess accurately the head injury potential of the mixture of materials that make up the impact surface.

Simulation results are represented graphically as a profile of acceleration on impact (*g*) over a period of time (*t*) known as a *g*/time waveform or trace (Fig. 10).

These *g*/time waveforms are used as the basis for measuring the severity of head injury on impact in head injury models (HIM).

3. Methods for assessment of head injury potential — head injury models (HIM)

3.1. The peak g method

The peak *g* method utilises the maximum recorded acceleration during an impact event (Fig. 11) and can provide information with regard to the ability of a surface to absorb the energy of impact. For a given fall height, a high peak *g* suggests that a surface provides little cushioning on impact, i.e. the object slows down over a short time period. A relatively low peak *g* suggests the surface cushions the impact, i.e. the object slows down over a longer time period and the surface absorbs a greater proportion of the impact energy.

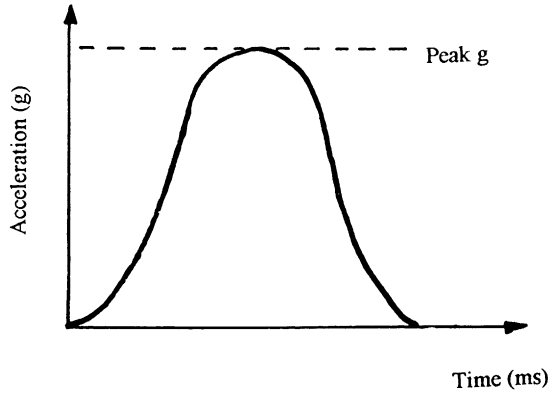

Fig. 11. Maximum acceleration on a *g*/time trace.

Gadd [18] commented that "the peak *g* is the most widely used rule-of-thumb measure of injury hazard because it is the simplest". He did however express reservations about the application of the method, recognising limitations when interpreting a single point on an acceleration/time waveform (von Gierke [20]). The peak *g* method utilises only the maximum acceleration (*g*) value — the duration of the impact pulse is not considered.

For example, the following acceleration time profiles give possible simulation results for two different impacts (Fig. 12). It can be seen that the peak accelerations are the same but the durations differ. The shorter duration impact may cause ASDH; an ASDH is unlikely during the longer duration impact [21]. If the peak *g* criterion were used as a head injury measure to assess the severity of these two impacts, it would give the impression that the same type of head injury could be expected from both.

A study of the current literature reveals a division of opinion concerning the peak *g* a child could be subjected to without sustaining a serious head injury.

The National Bureau of Standards study conducted by Mahajan and Beine [22] stated that Mohan et al. [23] "reported that a conservative estimate of head injury tolerance limits for head-first falls of children are (150–200)*g* average acceleration for 3 ms, or (200–250)*g* peak acceleration". These estimates were made by simulating falls of six children (1–10 years old) and one adult (21 years old) using the Motor Vehicle Manufacturers' Association (MVMA) Two-dimensional Crash Victim Simulator computer model'. This figure was similar to that estimated by Hodgson et al. [24]. Mahajan and Beine [22] comment that "these data suggest that the risk of serious head injury due to head-first fall is minimal when the peak acceleration imparted to the head is 200*g* or less".

Reichelderfer et al. [25], referring to studies performed by The Franklin Institute Research Laboratories in Philadelphia at the behest of the Consumer Product Safety Commission (CPSC), suggested that "the maximal acceptable impact level was 50*g*; beyond this, serious injury begins to occur when a child's [age not specified] head is dropped in a free-fall". In an accompanying commentary, Sweeney [26] referred to "the fatal impact threshold of 160*g*" but did not cite the sources from which that figure was derived.

As recently as 1993 Lewis et al. [27] suggested "peak accelerations less than 200*g* are felt to be unlikely to cause a life threatening head injury". Bond and Peck [28] commented that the severity of childhood (age not specified) head

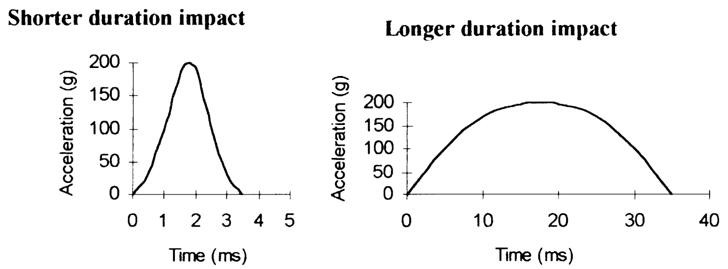

Fig. 12. Examples of acceleration time profiles from two different impact simulations.

injury due to falls in playgrounds is associated with the surfacing material at the point of impact suggesting "the maximum acceptable force is 50*g*; an impact force of 200*g* will deal a fatal blow".

King and Ball [6] felt that the general view coincided with that expressed by the Technischen Uberwachungs-Vereine (TUV) [29] that at values between 0 and 50*g* "one can be fairly certain of no permanent head injury damage to children" (age not specified). "At levels of from 50 to 150*g* the risk is small; from 150 to 200*g* there is a definite risk; and above 200*g* a grave risk".

Some of the previous extracts from the literature propose peak *g* limits for the assessment of the safety of children's playground surfacing without stipulating an age range. However, as the limit is for assessing playgrounds it can be assumed that these values are suggested as a measure of head injury potential in all children regardless of age (so long as they are old enough to be capable of playing there).

Although the wording is sometimes ambiguous, these extracts from the literature suggest agreement that 50*g* is the maximum acceptable limit before-injury threshold and that 200*g* is the threshold for fatal injury. This can be seen on the Wayne state tolerance curve (WSTC).

3.2. The Wayne state tolerance curve (WSTC)

Experimental work by Lissner et al. [30] at the Wayne State University in Michigan involving drop tests of four human cadavers with measurement, and comparison, of acceleration, intracranial pressure and structural damage laid the foundations for what is now known as WSTC.

The WSTC (Fig. 13) demonstrated that the severity of head injury was dependent both on the magnitude and the duration of impact.

Values above the curve suggest a 'danger to life' and values below are 'tolerable'. For example, from the WSTC shown in Fig. 14 it can be seen that an impact force of 100*g*

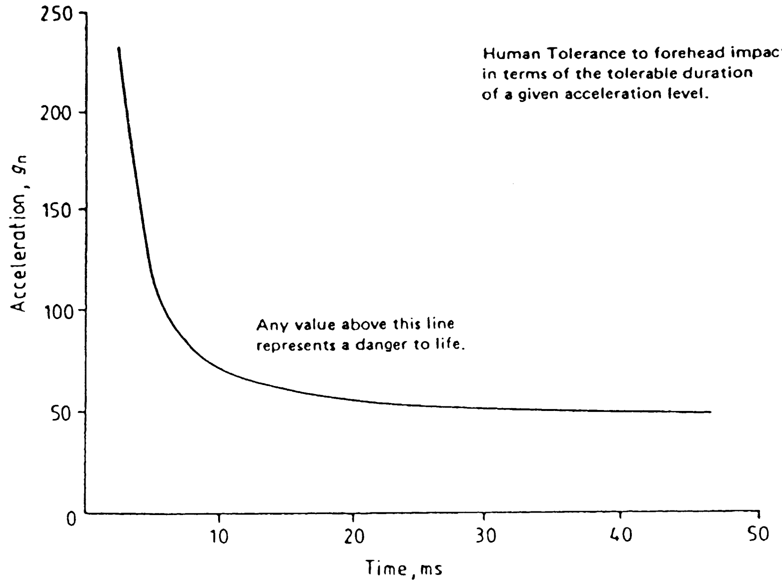

Fig. 13. The WSTC [31].

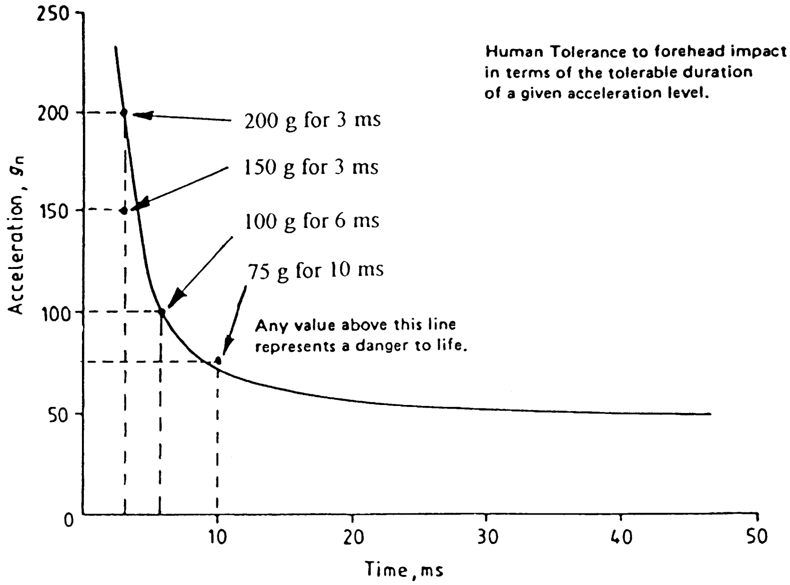

Fig. 14. Using The Wayne State University tolerance curve.

for 6 ms represents the same threshold of injury as 200*g* for 3 ms as they both lie on the curve. However, an impact force of 75*g* for 10 ms would represent danger to life but 150*g* for 3 ms would be tolerable.

The WSTC suggests that very intense head accelerations are tolerable if they are brief.

Versace [32] suggested that the WSTC "tolerance limit is probably not accurate. The impact data on which it is based were sparse and they were not very representative. It resulted from animal tests involving frontal hammer blows and air blasts to the exposed brain and from drop tests of human cadaver heads".

Collantes [9] commented that "The use of cadavers requires investigators to compensate for the differences in elasticity and compressibility of a living human head" and that "furthermore this linear model does not address the issue of brain damage due to angular acceleration. Studies have shown that angular acceleration rather than linear acceleration produces brain injury" citing numerous references to support those criticisms [33–39].

As discussed previously (Figs. 7 and 8), in addition to linear acceleration, angular and axial rotational accelerations may act on the head. Where the acceleration is predominantly angular or axial rotational, the application of a linear HIM alone may not be appropriate and other types of HIMs should be considered [38].

Although the WSTC overcame the problem of the peak *g* method by considering both the magnitude and duration of impact, Versace [32] was concerned that "It is difficult to know where to plot a point on a graph of acceleration versus duration if the rules for determining the numerical value of the acceleration are vague or ambiguous. An even more fundamental defect ... is the absence of a measure for acceleration level which is consistent regardless of waveform, the 'effective acceleration' ".

Collantes [9] was concerned further, regarding the use of a single value that is 'effective acceleration', suggesting in 1990 that "The WSTC attempts to show a relationship between linear acceleration, time, and severity of injury. A major flaw with this approach is that acceleration can only assume a discrete (single) value for any given duration as depicted on the curve. In reality, acceleration rises and falls during the period of impact", a criticism voiced earlier by the National Playing Fields Association (NPFA) which suggested (of a *g*/time curve) that "no single point on that curve can adequately describe the severity of that particular impact" [40]. The acceleration forces acting on a human head change continuously from zero, to peak *g*, to zero during the total duration of impact. During the first section of impact, the surface and the head will be subjected to compression whilst the head decelerates; the second section involves a small period of time where the head is almost stationary between deceleration and acceleration while the third section involves the deformed surface/skull/scalp springing back to its normal shape, and the head rebounding off the surface. This can be illustrated on a typical acceleration/time trace (Fig. 15).

However, Collantes [9] suggested that "although the WSTC was derived from a very small test sample, it does provide a starting point for determining the tolerance levels for head impact within the test parameters and for developing criteria to evaluate the different levels of injury". The authors are not aware of any attempts to adjust the WSTC for use with children.

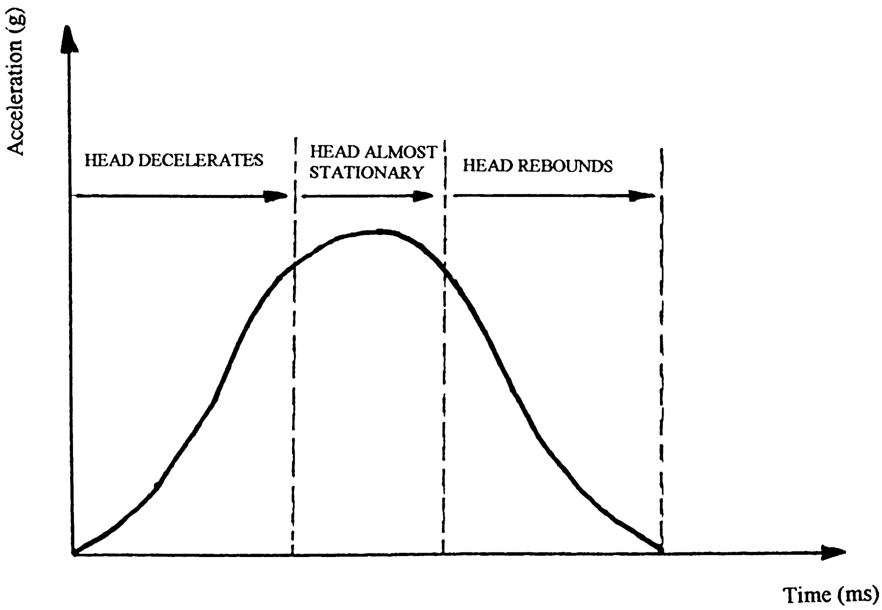

Fig. 15. Typical g/time trace of a human head during impact.

3.3. The Gadd severity index (GSI)

The Gadd severity index (GSI) [18] was developed in 1966 by Charles Gadd for correlating the likelihood of injury with impact during laboratory testing of automotive interior structures and components, thus providing indices with which to draw comparisons between alternative designs for improved safety [18].

The GSI utilised the WSTC data and produced an improved measure of head impact injury, which considered the pulse waveform in its entirety rather than just one aspect or point, acknowledging that injury is a function of both intensity of the loading and its time duration.

In addition, Gadd recognised that the diverse nature of an impact event involving different sites of injury, different angles of impact and different mechanisms of injury would make it "unrealistic to assume that we will ever have a single and rigorously quantitative rating system for the hazard inherent in a given pulse applied to a given part of the human body". Gadd did however concede the requirement for "simple measures or at least 'yardsticks' " for correlating injury with impulsive loading [18].

Gadd was able to produce a relatively simple formula for the derivation of indices for the prediction of head injury survivability. A simple exponential weighting factor (2.5) was applied to the integral of the acceleration–time pulse waveform, thus allowing an expression of a single number, which could be compared with an injury scale.

The GSI is defined as

$$I = \int_0^t a^n \, dt < 1000$$

where I is the function of acceleration and time of impact, n the weighting factor based on previous experimental data, a the acceleration on impact (in units of gravity, g), dt the duration of acceleration on impact (s), and 1000 is the threshold of danger to life for internal head injury during frontal blows [18].

The value for the weighting factor, 2.5, was suggested by Gadd [18] based on the gradient of the WSTC. Gadd suggested that "The exponential weighting may be thought of as recognising that the lower portions of the pulse contribute very little to the injury, but that the more intensive portions contribute to a disproportionately great degree".

The tolerance limit of 1000 was checked by Gadd [18] against various biomechanical data, including actual accidents simulated in the laboratory. The injury number of 1000 was considered to be the median point between those occupants who survived and those who did not survive. It was, however, not possible to exclude that those who did not survive received additional injury over and above that from frontal head impact alone. Therefore, an index of 1000 could indicate a survival limit of over 50%. Gadd indeed recognised that as further information becomes available, from research and survival studies, the threshold tolerance value of 1000 may require adjustment [18].

The GSI was designed for use in the automotive industry to compare the safety of various interior structures in vehicles. The most common head injury in automotive accidents is diffuse brain injury (DBI), Gennarelli and Thibault's study [21] of clinical cases suggests that "in patients without mass lesions unconscious for more than 24 h (diffuse brain injury), 89% were motor vehicle-related and only 10% were due to falls and assaults".

Hardy et al. [41] note in their literature review of head injury biomechanics — apparently quoting from Ommaya [42] — that "diffuse axonal injury (DAI) is more likely to occur under distributed loading conditions with longer duration (>10 ms) soft impact with negligible contact phenomena. Rotation is seen as a predominant factor. In cases of focused load, short duration (<10 ms) hard impacts where contact phenomena and translation are primary factors, DAI is less likely to occur". When the head strikes the dashboard in a vehicle collision the interior structures are designed to deform to reduce injury therefore increasing the stopping distance and subsequently the time duration of the impact. These longer duration soft impacts increase the probability of the occupant sustaining DBI.

The source data used for the GSI were based on tolerance to severe concussion [18], i.e. diffuse brain injury (DBI) rather than acute subdural haematoma (ASDH). Therefore, the GSI is applicable when trying to ascertain whether a vehicle's interior structure would cause life-threatening DBI; problems may arise when the GSI is applied to determining if a fall or assault could have caused a life-threatening brain injury. The most likely significant brain injury in a fall or assault is ASDH [21] — "the cause of the injury was by fall or assault in 72% of the ASDH group while only 24% of ASDH were due to motor vehicle-related injuries" — although analyses suggest that DAI may be a universal factor in fatal closed head injury [43] and that ASDH may be only a concomitant finding [44].

Head impacts due to falls and assaults typically involve higher acceleration rates than head impact due to vehicle collisions because of the unyielding nature of concrete and tarmac in falls, with or without preceding assault. ASDH is associated with a high impact force applied over a short time period. Based on animal studies and mathematical modelling, Gennarelli and Thibault [21] suggest that at "200*g* of acceleration with a pulse duration of 3.5 ms ... the acceleration strain-rate conditions are appropriate to rupture the bridging veins and cause ASDH".

The GSI value is calculated by integrating the whole waveform. Therefore, a relatively long duration, low acceleration impact could give the same GSI value as a short duration, high acceleration impact (Fig. 16). Impact tests on two different surfaces could give similar GSI values although the probability of ASDH rather than DBI is very different: it is difficult to know how precise a correlation is possible between these GSI values and the specific intracranial pathology.

The GSI assesses injuries to the head resulting only from impacts to the forehead. Gadd [18] suggested that "the most reliable information obtained is that for impact of the front of the head, and in particular the forehead" and commented, of the GSI, that "integration of this expression yields a severity index which is applicable to a particular class of injury ...", this injury being internal injury to the head from frontal blows.

The GSI has been applied to assessing the severity of head injuries in children. A method was developed by the British Standards Institution (BSI) to assess the heights at which equipment should be erected on a playground surface. A fall height where the GSI was equal to 1000 was the 'critical fall height' (CFH). The definition of the CFH was "the greatest height of head first fall from which a child, landing on a surface could be expected to avoid sustaining critical head injury" [31]. The age range to which the method applied was not specified. However, as the method was designed to assess playground surfacing it can be assumed that the method suggested is for use with all children old enough to be capable of playing there regardless of age.

Assessing internal head injuries resulting from frontal impacts would be applicable in a car crash scenario as the head would be thrown forward hitting the dashboard or steering wheel on the forehead or facial region (Fig. 17). The need for caution in the application of the GSI to other circumstances of impact-acceleration to the head has been emphasised [2].

Objections have been raised also to use of the GSI in the context of traffic safety: "The GSI was used as a standard for the first time in 1971 by the US National Highway Traffic Safety Administration (NHTSA). As a result of objections to GSI, NHTSA later adopted, starting with Versace's criticisms in 1971 [32], a new expression for head impact tolerance specification, the head injury criterion (HIC)" [46].

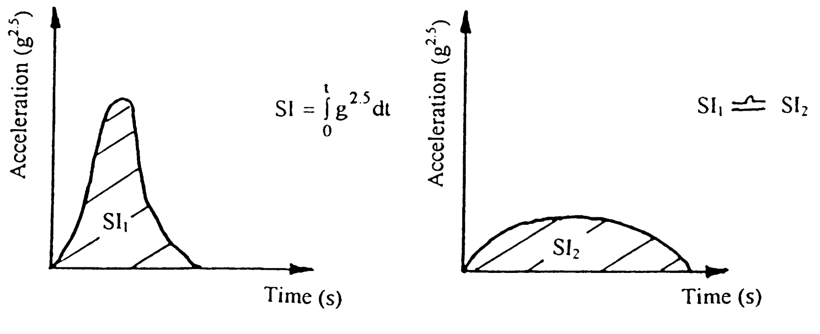

Fig. 16. Two different waveforms with similar GSI values

Fig. 17. Head-steering wheel hub contact in medium-hard contact impact [45].

3.4. The head injury criterion (HIC)

The HIC, first proposed by Versace [32] in 1971, is an alternative interpretation of the WSTC combined with information derived from experiments conducted on crash dummies [9,47].

The HIC addresses the shortcomings of the GSI, providing comparable head injury tolerance values irrespective of the waveform shape (Fig. 16). The HIC considers the more injurious portion of the impact waveform, the peak and close to peak sections (i.e. $t_2 - t_1$, Fig. 18), and excludes the less injurious sections therefore giving a more accurate head injury tolerance level.

The HIC is defined [9] as

$$\mathrm{HIC} = \left[(t_2 - t_1) \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a \, \mathrm{d}t \right\}^{5/2} \right]_{\max} < 1000$$

where $t_2 - t_1$ is the portion of waveform to be measured during which HIC attains a maximum value (Fig. 18), a the acceleration on impact (in units of gravity, g), and dt the duration of acceleration on impact (s).

Prasad and Mertz conducted a study in 1985, which involved the collection of skull fracture and brain injury data and their corresponding HIC values. The study concluded that "at an HIC of 1000 there is a 16% risk of life threatening brain injury" [48]. The HIC value is defined as the HIC tolerance threshold or critical limit/value.

Snyder et al. [49] believe the HIC to correlate well with observed injuries.

The integration time period is subjective and Hardy et al. [41] comment "there is a dispute regarding the duration of integration (i.e. $t_2 - t_1$) between the NHTSA and the International Standards Organisation (ISO). The NHTSA limit is 36 ms while the ISO recommended limit is 15 ms". Prasad and Mertz suggest that "HIC duration should be limited to 15 ms or less for the calculation of the HIC value for a given resultant head acceleration–time history ... for example, an average acceleration of 1g applied for 1000 seconds gives an HIC of 1000" [48] which would clearly not be life threatening. The NHTSA have also come to this conclusion, suggesting in 1999 [50] that the HIC time duration should be reduced to 15 ms. King and Ball [6] citing Hodgson and Thomas [51], also commented on this point in their study concentrating on childhood head injuries in playgrounds, stating "the HIC time interval ($t_2 - t_1$) would need to be less than 15 ms to pose a concussion hazard (which it generally is in playground type impacts), even were the HIC value to exceed 1000".

Prasad and Mertz [48] developed an injury risk curve from adult data (shown in revised form in Fig. 19). They found that the brain injury and skull fracture curves were virtually identical and stated that "the implication is that for a given level of HIC, skull fracture, brain damage or both are equally likely to occur" [48].

The probability of fatal head injury can be given for any HIC value between 500 and 2500, this injury risk curve could be useful for interpreting the probability of fatal head injury for a given HIC value. For example, a statement that

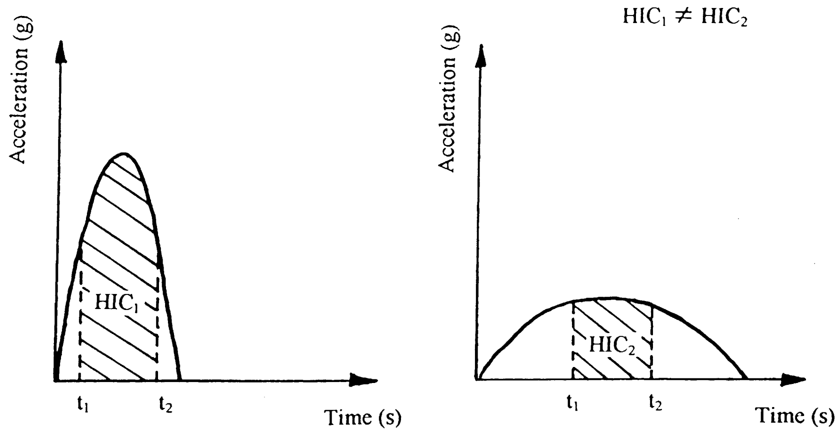

Fig. 18. Area of waveform utilised for calculating HIC [9].

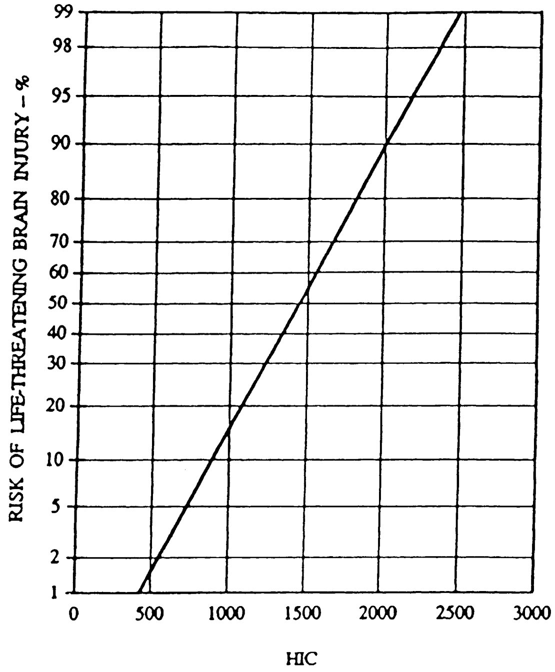

Fig. 19. Injury risk curve for HIC [52].

"an adult has a 20% probability of sustaining life threatening brain injury from the given fall scenario" conveys more to a lay person than "the fall scenario produced an HIC value of 1100".

Goldsmith [34] expressed concern regarding the HIC including "since data from which present stipulated criteria were derived are primarily based on the behaviour of adult cadaver skulls, it is specious to propose the same head injury criteria for adults and children, and this thesis has been well documented in a recent comprehensive article (Stürtz [53]) . . . furthermore, the criterion does not address the question of brain damage due to rotational acceleration [35]".

Stürtz [53] suggested the use of a 'critical load value' defined as "the load on the body under which an initial considerable damage of the organism takes place — destruction of a cell; irreversible injury — for instance when bone fractures occur or primary organs rupture". The author simulated 10 pedestrian accidents involving children (age not specified) with an anthropometric dummy. HIC values were correlated with accidents causing "the lowest irreversible injury severity degree, i.e. AIS 2 [abbreviated injury scale] . . . (e.g. skull brain trauma of first degree, SBT 1)" [53]. Stürtz suggested an HIC of 840 as the critical load value of a child's head.

The HIC is a linear head injury model designed for use in the automotive industry to assess impacts of crash test dummy heads onto the interior structures of motor vehicles. During crashes, in addition to linear accelerations (e.g. impact onto the dashboard), an occupant's head may be subjected to axial rotational and/or angular accelerations (e.g. whiplash). A failure to consider these accelerations may result in an under representation of head injury potential. However, in the context of a fall scenario where the impact is head-first the linear acceleration predominates.

Mertz and Weber [54] developed an injury risk curve applying to children rather than adults (Fig. 20). This study analysed the impact responses of a "3-year-old child

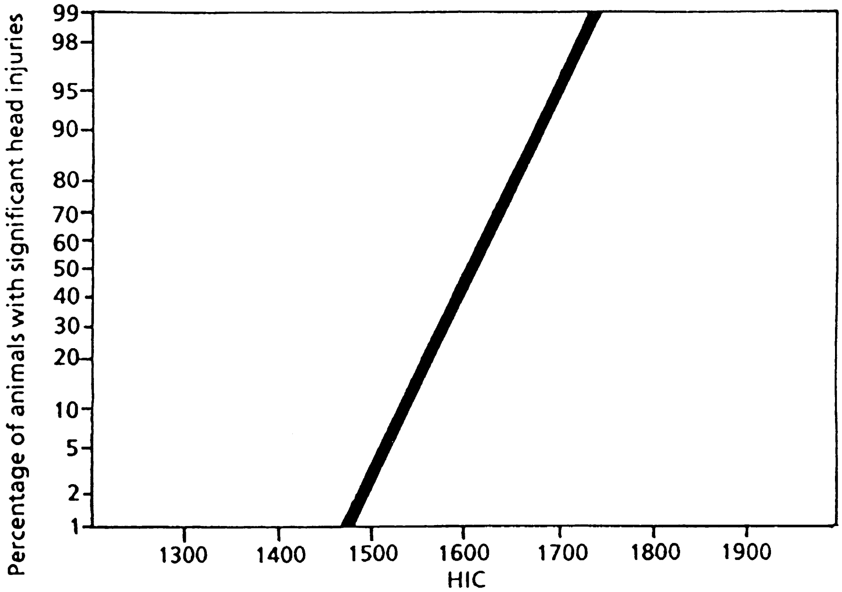

Fig. 20. Percentage of animals experiencing a significant head injury as a function of HIC experienced by a 3-year-old child dummy [54].

dummy'' and subjected anaesthetised animals (3-year-old child surrogates) to the same impact conditions. In this way a risk curve was developed giving the percentage of animals experiencing a significant head injury at the HIC value recorded by the child dummy under the same conditions. Significant head injury was defined as TL (threat-to-life) = 3, 4, 5 and 6. A TL equal to 0 was defined as no head injury and a TL equal to 6 was defined as non-survivable head injury.

Federal Motor Vehicle Safety Standard 208 [55] for frontal crash protection and the European Enhanced Vehicle-Safety Committee (EEVC) Working Group 17 Report [56] for evaluating pedestrian protection, currently require the use of the HIC for assessing the severity of head impact scenarios. It is also specified as the HIM in American Standards for Testing Materials (ASTM) F 1292–93 [57] and British Standards Institution (BSI) ENBS 1177 [58] for assessing the severity of head first fall/impact simulations onto various surfaces. BSI comment that the HIC ''is considered to be the best model available to predict the likelihood of injuries from falls'' [58].

4. Conclusions

Information regarding the tolerance of the human head to impact originates from various sources including experimentation on animals, human cadavers, anthropomorphic dummies and human volunteers.

Inaccuracies can arise from using such data: skulls of animals and humans are different in shape and thickness; mechanical properties such as compressibility, elasticity, pressure and density vary between the living and the dead. Anthropomorphic dummies do not replicate the complex biological system of the living human body; human volunteers can only be subjected to tests involving non-injurious accelerations. Correlation between clinical presentation and mortality may differ between children aged 14 years or less and adults [59], as may the spectrum of brain injury [60].

In spite of these limitations, it is desirable that reliable predictive head impact indices are developed to assist the design of motor vehicles, safety helmets and playground surfaces and to aid the investigation of causation of head injury by forensic bioengineering. The peak *g*, WSTC, GSI and HIC have previously been utilised in assessing the severity of head injury from simulated playground falls, motor vehicle crashes and forensic investigations into childhood falls. Current playground surface testing standards for the USA [57] and the UK [58] stipulate the use of one or more of these models. Motor vehicle crash test standards [55] and pedestrian safety recommendations [56] stipulate the use of the HIC; published reports of forensic investigations have utilised the peak *g* [1] and GSI [2]. The GSI and HIC were developed utilising data from which the WSTC was derived. Other HIMs exist — mean strain criterion (MSC), Japan head tolerance curve (JHTC), brain compliance model (BCM), to name only three discussed in detail by Hardy et al. [41] and McElhaney et al. [61] — but have not been applied to the circumstances under review here. There are many theories and suggestions for future HIMs in the literature but these have yet to be accepted by the government bodies. NHTSA suggested in their 1999 document on improving injury criteria that ''while considerable progress has been made in the capabilities of analytical finite element head/brain models to simulate the major injury mechanisms prevalent in brain injury, it was felt that it would be premature for their results to be used in this current proposed rulemaking action'' [50].

The HIC is widely accepted to be the current 'state-of-the-art' in head injury models and should be used in preference to, or to corroborate, such other models as the peak *g*, WSTC, and GSI. The use of such models may at least allow a prediction that the injury is or is not likely to have been produced by the event described in the case history (provided that the event can be 'simulated' to allow quantification of those parameters required for the models). Although theoretical analysis of impacts (using the methods and formulas described in the first part of this paper) can be useful in approximating the severity of a head injury from an alleged fall, an in-situ simulation has been shown [19] to be more accurate in assessing the head injury potential of impacts onto surface mixtures.

If HIMs are used to obtain quantitative information in cases of possible child abuse it is important that they are implemented with caution, proper emphasis being placed upon relative advantages and disadvantages, and that these limitations are specified. The resulting objective data which can be presented to the court may either substantiate or refute a given head injury scenario.

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