SLOVENSKI STANDARD
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Merila za določanje poškodb glave, ki nastanejo zaradi udarcev in opeklin - Vodilo za odbore CEN, ki pripravljajo standarde za čelade

Head impact and burn injury criteria - A Guide for CEN helmet standards committees

Kriterien für Verletzungen durch Einwirkung auf Kopf und Verbrennungen - Leitfaden für Arbeitsgruppen, die europäische Helmnormen erarbeiten

Critères relatifs au traumatisme crâniens et aux brûlures - Guide destiné aux comités des normes sur les casques de protection du CEN

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Head impact and burn injury criteria - A Guide for CEN helmet standards committees

Critères relatifs au traumatisme crâniens et aux brûlures - Guide destiné aux comités des normes sur les casques de protection du CEN

Kriterien für Verletzungen durch Einwirkung auf Kopf und Verbrennungen - Leitfaden für Arbeitsgruppen, die europäische Helmnormen erarbeiten

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COMITÉ EUROPÉEN DE NORMALISATION
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European foreword

This document (FprCEN/TR 16148:2026) has been prepared by Technical Committee CEN/TC 158 'Head protection'

This document is currently submitted to the Vote on TR.

This document will supersede CEN/TR 16148:2011

In comparison with the previous edition, the following technical modifications have been made:

— added Clause 1 Scope and Clause 3 Terms and definitions;

— added Table 1 with test methods and metrics correlated to different injury risks;

— updated Table 2 AIS scale with head injury severity;

— differentiating between linear and rotational shock absorption;

— updated metrics for linear shock absorption;

— added metrics for rotational shock absorption;

— added 5.4 Force transmission;

— added Clause 8 on Finite Element (FE) based metrics;

— removed Neck injury as it is not within the scope of head protection;

— removed Noise injury as it is not within the scope of head protection.

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Introduction

Members of helmet Standards committees frequently need to choose a test procedure and define limits for the test procedure. Such limits should represent test values that indicate the potential for injury corresponding to the accepted risk level. Over the years, injury criteria, or risk functions have been developed for different body regions derived from a combination of accident and casualty data, as well as data from tests on cadavers, animals or human volunteers. It is important to stress that a test response should always be measured as a risk of injury. The human response to a given dose of applied load varies across a range of the population and not all humans would respond similarly to the same load.

It should further be noted that the available risk functions all have their limitations. The available data consist only of subgroups representing a fraction of human data, or they are scaled from animal or cadaver data, which introduces uncertainties regarding their accuracy. However, such injury criteria are often used by the industry as pass/fail values without considering the variation in human tolerance to injurious forces. This sometimes leads to the mistaken belief that any value below the stated limit implies uninjured and all values above imply a serious or fatal injury. Instead it should be concluded that the specified limit values are intended to ensure an adequate level of safety with a low risk of injury for the assessed impact energies.

Another aspect in injury mechanics to the head is that many injury risk functions are developed by the automotive industry and thus represent typical automotive accidents. The head kinematics in automotive accidents and helmeted head impacts are different. This means that the kinematic metrics developed for automotive application as well as their thresholds, is not suitable for assessing helmets.

Many accidents to wearers of helmets, which cover a wide range of activities from sports like horse riding or downhill skiing to industrial work and fire fighting, result in a closed head injury. This is when the brain is damaged without any skull or external tissue damage. Historically, the main concern in head injury prevention has been skull fractures and the test methods have focused on assessing pure translational loads, mainly connected to skull fracture. This report introduces available metrics related to brain injuries produced by rotational loading that can be used in helmet standards to assess the risk of closed head injury as well as skull fractures.

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Children and adults have distinct pathophysiology and injury patterns, indicated by a meta-analysis of 48,074 child cyclist head injuries, suggesting their helmets should not be tested the same way [1]. Two factors are expected to affect the protection level of helmets, the characteristic of the head impact and the head injury tolerance. The real-world head impact biomechanics of children are poorly characterised. Unlike in adults [2], we have less or no quantified accounts of common head impact speeds, angles, or locations in child cyclists, so it is unclear how closely the conditions used in current child helmet test methods represent real-world incidents. Nevertheless, given differences in travel speeds, fall heights, and vehicle wrap-around geometry, child impact characteristics are likely to be distinct from those of adults [3]. The second factor is that children are expected to have a lower head injury tolerance than adults. Their proportionally larger heads and weaker cervical musculature amplify head excursion and angular kinematics under a given loading [4], [5], their thinner and more deformable skulls alter contact mechanics at equivalent loads [6], and their lower-mass heads rotate more under a given load, producing greater brain tissue strain[7]. Together, these factors suggest the need for helmet test conditions and pass/fail limits tailored to children, but further work is needed to determine child-representative impact conditions and injury-driven pass thresholds.

The intention of this document is to supply the members of helmet standards committees with a summary of available injury criteria and metrics for a range of injuries and risk levels, corresponding to available test methods. This revised version of the report has increased the focus on the relation between test method and injury metrics, included metrics related to closed head injury and has removed the chapter on neck injury to focus on assessment of head protection.

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1 Scope

This document outlines criteria related to head injury risk and burn injury risk in the context of helmet testing. It is intended to support the work of CEN/TC 158 working groups by providing guidance on selecting appropriate test methods and defining injury thresholds based on injury type and severity.

This document does not include requirements or test methods.

2 Normative references

There are no normative references in this document.

3 Terms and definitions

No terms and definitions are listed in this document.

ISO and IEC maintain terminology databases for use in standardization at the following addresses:

- ISO Online browsing platform: available at http://www.iso.org/obp
- IEC Electropedia: available at http://www.electropedia.org/

4 Injury prevention and testing

4.1 General

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A head injury caused by an impact is usually referred to as a traumatic head injury. The severity of the injury depends on the load case and what protection was used. The purpose of the helmet certification is to measure the protective performance of the helmet in a specific load case representing a set of accident scenarios. The test method also needs to be robust and repeatable. For each chosen test method there is a corresponding injury metric that can be measured and analysed if the helmet performs according within given threshold values. If it exceeds the thresholds the helmet does not fulfil protection requirements according to standard. The injury metrics are chosen to cover the most relevant types of head injuries.

Head injury is a broad term that covers a wide range of injuries to the scalp, skull, brain and underlying tissues and blood vessels in the head. Broadly, these injuries can be categorised as superficial injury, skull fracture or brain injury as defined by what tissues are injured, see Figure 1. It can also be a combination of any of the three where e.g. a fracture causes damage to the brain. A more detailed description of head anatomy and different types of injuries is found in the Annex A.

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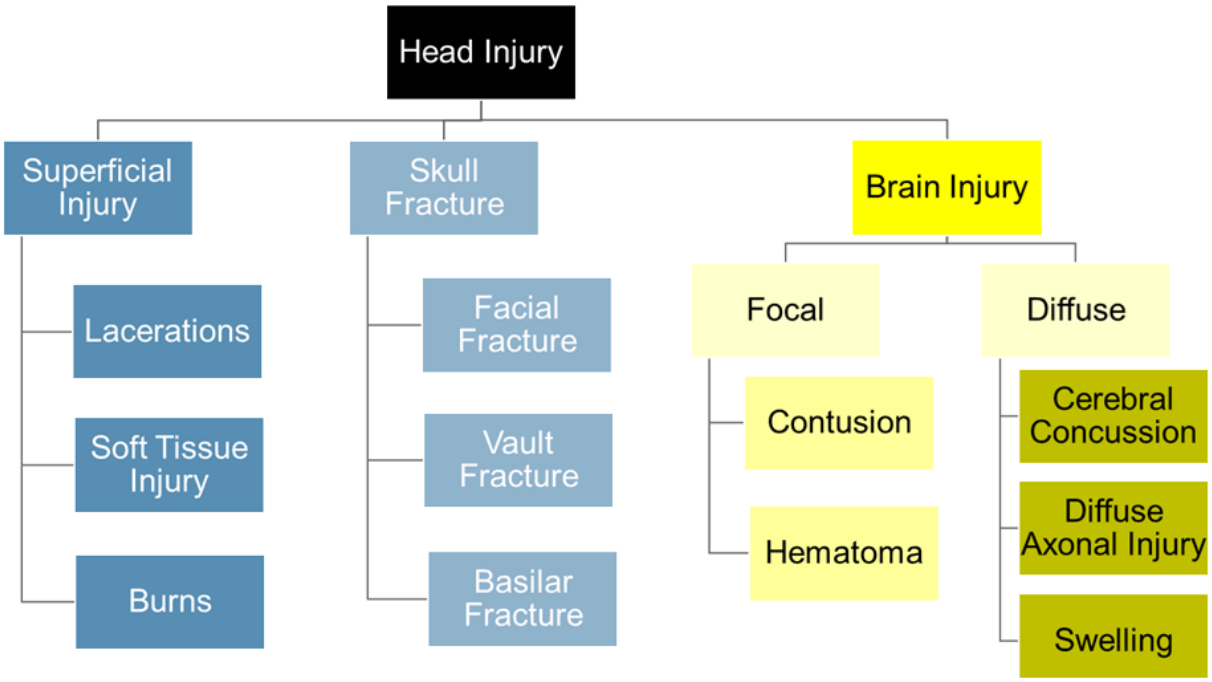


Figure 1 — Head injury types

4.2 Injury metrics

Helmet standards are designed to ensure that helmets provide sufficient protection in relevant accident scenarios and help reduce the risk of associated injuries. Since the injury mechanism is different for different tissues so e.g. the skull and brain are sensitive to different types of loading, helmets needs to be tested in various test set-ups and assessed with injury specific metrics.

Several studies have shown that for helmeted riders, cerebral concussion is the most common head injury [8][9]. Also, previous studies have shown a larger reduction in skull fractures compared to brain injuries when wearing a helmet compared to not wearing a helmet [10][11]. However, both skull fractures and brain injuries occur in helmeted bicycle accidents [8]. Also, within construction it has been shown that concussion and internal injuries are more prevalent than fractures among reported head injuries [12]. Previous research has shown that many of the brain injuries correlate better to rotational kinematics [13][14][15] whereas skull fractures better correlate with linear kinematics [16][17].

Many different injury metrics have been proposed over the years. Some are solely based on linear kinematics such as Head Injury Criterion (HIC) 5.3, some solely on angular kinematics such as Brain Injury Criterion (BrIC) 6.4, whereas some have a combination of linear and angular kinematics, e.g. Head Acceleration Response Metric (HARM) 7.2. There are also injury metrics that are based on the response from Finite Element (FE) models, such as maximum principal strain and von Mises stress 8.1. At the moment there is no consensus about the most suitable injury metrics to predict different types of head injuries but many are used in parallel.

There are a number of different test methods described in various standards. These are used interchangeably or in combination when they measure different properties of the helmet. This report includes a description of the injury metrics related to the most common test methods in the European helmet standards.

Different test methods evaluate different loading conditions, and the metrics represent risks for different injuries. See Table 1. It should further be noted that the test methods use different headforms depending on what injury risk or metric it is intended to represent. The headform properties are specific for the requirements of the test and cannot be interchanged with any other test headform without affecting the results negatively.

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Table 1 — Test methods and injury metrics associated with risk

Risk type	Blunt impact from — Falling objects — Falls		Crush	Penetration — Sharp impact — Projectile
Typical associated injury	Skull fracture Contusion Epidural haematoma	Concussion Focal injury e.g. subdural haematoma Diffuse axonal injury	Skull crushing	Skull penetrating
Protective property	Linear shock absorption	Rotational shock absorption	Structural rigidity	Penetration protection
Test methods	Dropped headform Striking object	Dropped headform	Lateral deformation	Penetration
Measured metric	Force or Linear acceleration	Rotational acceleration or Rotational velocity	Magnitude of deformation	perforation or not
Global injury criteria	PLA HIC Peak transmitted force SFC	PRA, PRV, BrIC Combined metrics	N/A	N/A
Tissue based injury criteria	Internal energy (FE)	Strain (FE) Stress (FE)		

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4.3 Injury risk assessment

When assessing injury risk in biological tissues, variations in human anatomy and tissue properties across the population are to be considered. Individuals differ in their sensitivity to impact forces, resulting in varying probabilities of injury under the same loading conditions. While most people may tolerate a certain level of strain, some may sustain injuries at that same level due to lower tolerance. Consequently, there is no absolute threshold that definitively separates injury from non-injury. Instead, injury risk is represented by risk functions - a dose response curve - which express the probability of injury as a function of a load parameter, such as force or strain. Risk functions are typically derived from injury statistics and experimental data and expressed with statistical distributions.

The dose response curve tends to be "S" (sigmoid) shaped such that as the magnitude of the injurious parameter increases so does the percent of the population that sustains an injury of a given severity. Thus, a family of "S" curves can be generated for a range of injury severity such as Abbreviated Injury Scale (AIS) 4.4 and a measurement or criterion such as HIC 5.3. Unfortunately, the data needed for generating risk curves is generally difficult to obtain. Injury statistics does not give the exact level of load and measurements generated by test apparatus do not relate directly to injury severity since surrogate headforms and instrumentations does not respond like a human head in an impact. Hence, it is necessary to find a relationship between these test measurements and injury severity.

Risk functions usually represent a subgroup of the entire population, such as American football players, and thus the risk is biased based on that subgroup's characteristics. Many values used for injury risk analysis represents a 50th percentile male, which means that male of average stature would sustain an injury of a given AIS severity for the corresponding range of acceleration.

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Furthermore, in real life, the helmet impact scenario will vary from minor impacts to very high loading conditions as well as cover many different surface conditions and impact angles. In a helmet standard the number of tests performed are limited due to cost. Thus, for the standard test requirement, both impact severity and injury risk threshold are chosen as a representation of typical accident scenarios and injury mechanisms. The certification testing of head protection use pass/fail criteria to ensure that the helmet has a minimum level of protective capabilities. However it is also taking into account that it should be possible to design a helmet with reasonable dimensions, making it acceptable for the user to wear. Note that historically, risk values have been set which correspond to AIS 5 severe and that this has resulted in helmets that have given reasonable protection.

The risk and severity of the injury that may be tolerated in different helmet standards may vary depending on the chosen scenarios. This is up to the standard to define but could consider e.g.:

a) Minimising the risk of death and disability following a head impact - allowing moderate injury.

b) Reducing the risk of long-term brain injury, e.g. Chronic Traumatic Encephalopathy (CTE), from repetitive head impacts - protecting also for minor injury.

c) Lowering the likelihood of incapacitation following a head impact, thereby reducing use-case specific hazards such as drowning or post-crash fires - protecting against minor injury such as concussion.

4.4 Abbreviated Injury Scale, AIS

The Abbreviated Injury Scale (AIS) is a scale that extends from 0 to 6 where 0 is uninjured and 6 is unsurvivable [18]. Each level can be applied to anybody region according to a coding manual developed by the Association for the Advancement of Automotive Medicine (AAAM). Table 2 gives the AIS level for different types of head injuries based on AIS2015. Table 3 gives examples of typical head injuries for each severity level. Table 4 gives similar information for burn injuries by degree, surface area and region of the body.

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Table 2 — AIS level of different types of head injuries based on AIS2015

	AIS1	AIS2	AIS3	AIS4	AIS5	AIS6
Scalp	x	x	x			
Artery			x	x	x	x
Vein		x	x			
Brain stem					x	x
Cerebellum		x	x	x	x	
Cerebrum		x	x	x	x	
Brain swelling		x	x	x	x	
Pituitary			x			
Cerebral concussion		x	x			
Diffuse Axonal Injury				x	x	
Base fracture			x	x		
Vault fracture		x	x	x		
Headache	x					
Crush injury						x
Avulsive injury						x
Penetrating injury			x		x	x

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Table 3 — Examples of head injuries in the different AIS levels

AIS level	Examples of head injury
AIS1	Headache, scalp abrasion or minor laceration
AIS2	Cerebellum tiny subdural haematoma, mild concussion with 0 - <1 hours loss of consciousness, closed vault fracture
AIS3	Total scalp loss, cerebellum small to moderate subdural haematoma, mild brain swelling/ oedema, cerebrum tiny subdural haematoma, severe cerebral concussion with loss of consciousness 1-6 hours, closed base fracture, comminuted vault fracture
AIS4	Cerebrum small to moderate subdural haematoma, mild diffuse axonal injury with coma 6-24 hours, complex base fracture, complex or massive vault fracture
AIS5	Brain stem contusion, cerebrum large subdural haematoma, diffuse axonal injury with coma more than 24 hours
AIS6	Brain stem laceration, crush injury or massive destruction, avulsion injury

Table 4 — AIS scale with burn injury severity

	AIS 1 minor	**AIS 2 slight**	**AIS 3 moderate**	**AIS 4 serious**	**AIS 5 severe**	**AIS 6 unsurvivable**
1st degree	X	X				
unspecified	X					
2nd degree	X					
< 10 % TBS[a]	X					
3rd degree	X	X	X			
< 10 % TBS[a]		X				
< 10 % TBS[a] with face, hand or genitalia involvement			X			
2nd or 3rd degree						
10 % to 19 % TBS[a]		X				
10 % to 19 % TBS[a] with face, hand or genitalia involvement			X			
20% to 29% TBS[a]			X			
20% to 29% TBS[a] with face, hand or genitalia involvement				X		
30 % to 39 % TBS[a]				X		
30 % to 39 % TBS[a] with face, hand or genitalia involvement					X	
40 % to 89 % TBS[a]					X	
≥ 90 % TBS[a]						X
[a] Total body surface						

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5 Linear shock absorption

5.1 General

Linear shock absorption is primarily measuring the risk of skull fracture, contusions and epidural haematoma. The most common test methods are to drop a headform on an anvil and the inverted, to drop a striker onto a rigid headform. The linear shock absorption can be measured either in straight or angled impacts. Below are the most commonly used injury criteria and risk levels described.

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5.2 Peak Linear Acceleration (PLA)

This is the most frequently used parameter in helmet testing and is derived usually from a triaxial accelerometer mounted in the headform unless the headform is rigidly supported and then the source is a single axis accelerometer. In both types, the helmet is mounted onto the headform and then the apparatus allowed to fall unimpeded onto a rigid anvil. One study used Post Mortem Human Subject (PMHS) tests, FE modelling and drop tests using HIII headform to construct a risk function for skull fractures based on PLA [17]. This study suggests a 50 % risk of skull fractures for PLA = 220 g. This study used old subjects. A recent study reconstructed collisions between cars and vulnerable road users aged 60+ and constructed a PLA risk function for head injuries of AIS4+ [19]. They found 50 % risk of AIS4+ head injuries is PLA = 200 g. The injury threshold for the normal population is likely to be higher than these values. Probably PLA = 250 g, suggested by [20], provides a better estimation of 50 % risk of AIS4+.

A Finite Element (FE) study by Li et al. [21] evaluated risk of skull fracture for child head with varying age (1,5; 3; 6 and 18 years). They suggested that younger individuals should have a lower threshold for peak linear acceleration because they are more sensitive due to the fontanelles. They propose to have a critical value of 150 g for helmets for 1,5 years old or younger. This study is until now one of few studies evaluating the age-dependency in helmet testing and more data to set specific criteria for children are required, but the increased risk for children should be considered in standard writing.

5.3 Head Injury Criterion (HIC)

The Head Injury Criterion (HIC) [22] was developed from the Gadd Severity Index (GSI) and is based on the Wayne State tolerance curve [23]. The Wayne State Tolerance Curve was derived from a body of experimental data, including both skull fracture and brain injury data, where the short duration impacts were from skull fracture data. It evolved from the work of Lissner *et al.* [23], Gurdjian *et al.* [24][25] and Patrick *et al.* [26], and gives the tolerable average acceleration in A-P direction (Anterior-Posterior) as function of the pulse duration. HIC is calculated from the resultant linear acceleration of the head's centre of gravity over time [27]. Unlike peak linear acceleration, HIC accounts for the duration of the load. This is important because impacts with the same peak acceleration, but different pulse durations can lead to different injury outcomes. The HIC Formula (1) is shown below:

$$HIC = \left[\left(\frac{1}{t_2 - t_1} \cdot \int_{t_1}^{t_2} a_{res} dt\right)^{2,5} \cdot (t_2 - t_1)\right]_{max} \quad (1)$$

Where a_{res} is the resultant head acceleration as a function of time and t_1 and t_2 define the time interval over which the integral produces its maximum value. The duration t_2 - t_1 is set to either 15 ms or 36 ms, yielding HIC_{15} or HIC_{36}, respectively. In helmet impacts, the impact duration is typically less than 36 ms, making HIC_{15} the more appropriate metric.

Hodgson and Thomas [28] suggested that the critical HIC interval is less than 15 ms, even if the HIC value exceeded the threshold of 1 000 over a longer interval. His finding was based on examination of events where the concussive outcomes were known or could be determined.

Historically the threshold for injury "dangerous to life" was set as 1 000 for both GSI and HIC [22]. HIC 1000 was confirmed to correlate with the occurrence of bone fracture in a study by Hodgson [16] and reports by Mertz and Prasad et al [29][30] correlated HIC with the risk of skull fracture where HIC 700 corresponded to a 9-16 % risk of skull fracture, HIC 1 000 corresponded to 19-40 % , HIC 1 500 48-82 % and 2 000 66-90 % risk of skull fracture, the variation depending on the data set studied. A European project looking at motorcycle accident data [31] provided tentative values where an injury severity of AIS 2 was predicted for HIC 1 000 and AIS 3 at HIC 1 500, although, they did not correlate well. The limit presented by NHTSA for occupant safety is 700 for HIC_{15}[32].

5.4 Force transmission

A number of test methods use the transmitted force as a metric. That is the force transmitted from a striker impacting the helmet as measured in a force plate below the headform. Direct force can produce

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skull fractures and focal brain injuries. Tolerance to skull fracture depends on the location, contact area, and rate of the applied force, with tolerance ranging from less than 5,000 N to over 10 000 N [33].

5.5 Skull Fracture Correlate (SFC)

SFC was developed by Vander Vorst et al. [34] as an improved skull fracture predictor based on kinematics. It is defined according to Formula (2) where ΔV_{HIC} is the change in velocity over the time interval and ΔT_{HIC} is the time interval (t_1 and t_2) that maximizes the HIC value, usefully either maximum 15 ms or 36 ms known as HIC_{15} or HIC_{36}.

$$SFC = \Delta V_{HIC} / \Delta T_{HIC} = \int_{t_1}^{t_2} a \, \mathrm{d}\, t \Big/ (t_2 - t_1) \Big| \quad (2)$$

With logistic regression of experimental data of postmortem human subjects a risk curve was developed combining both frontal and lateral impacts [35]. According to that risk curve, the 15 % probability of skull fracture with SFC15 occurs at 124 g.

6 Rotational shock absorption

6.1 General

Rotational shock absorption asses the helmets protective performance regarding rotational motions. Injuries related to rotational motion are primarily diffuse brain injuries like concussion and DAI but also vascular injuries like subdural haematoma. To evaluate the rotational shock absorption the test method needs to allow for the headform to rotate, either freely or connected to a biofidelic neck.

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6.2 Peak Rotational Acceleration (PRA)

Peak Rotational Acceleration parameter is known to contribute substantially to brain injury but the relationship with injury is difficult to quantify and this relationship depends on the application. It is possible to measure rotational kinematics with instrumentations like a nine-accelerometer or rate sensors in the headform.

Depreitere et al. [36] evaluated the risk of acute subdural haematomas from bridging vein rupture. They performed experiments with post mortem human subjects by impacting them at the back of the head with a pendulum with varying energy and impact duration (4-15 ms). They concluded from the study a tolerance curve at around 10,000 rad/s^2 for pulses with a duration shorter than 10 ms.

Davidsson et al. [37] performed animal experiments with rats to evaluate the injury threshold for diffuse axonal injury. The loading was in the sagittal plane rearward. By scaling the rat experiments they found a threshold for DAI of 10 000 rad/s^2 with a duration of 4 ms.

Anderson et al. [38] presented a risk curve based on reconstruction of American football reconstructions with hybrid III dummies [39] [40]. They found a 50 % risk of sustaining a concussion at around 5 700 rad/s^2. Rowson et al. [41] also used data from American football but the data were taken from helmet mounted sensors. They found a 50 % risk of concussion at 6 383 rad/ s^2.

6.3 Peak Rotational Velocity (PRV)

Peak rotational velocity is another metric used for evaluating rotational induced brain injuries. Anderson et al. [38] presented an injury risk functions based on reconstructions from the national football league [39] [40]. They found a 50 % risk of mild traumatic brain injury at 38 rad/s. Meanwhile, Rowson et al. [41] presented a 50 % risk of a mild TBI (Traumatic Brain Injury) at 28 rad/s. Their risk function was based on mouthguard measurements from collegiate football players. Another study proposing a risk function for peak rotational velocity [42], where they used reconstructions of impacts in Australian football and rugby. They found a 50 % risk of concussion at 22 rad/s. Wu et al. [43] performed accident reconstructions of vulnerable road users (cyclists, pedestrians, electrical two

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wheelers) and found the 50 % risk of an AIS4 injury to be 27,8 rad/s. However, this risk function was developed from individuals that were 60 years or older, which could influence the risk since they can be more vulnerable to certain types of injuries such as subdural haematomas.

Margulies and Thibault [44] focused on diffuse axonal injury associated with prolonged coma and structural damage to axons. They used data from experiments on baboons (53 animals), analytical modelling and physical models of the human and baboon heads filled with a brain like material. The determined asymptotes for peak rotational acceleration and peak change in rotational velocity were 16 krad/s^2 and 46,5 rad/s for DAI AIS4+.

6.4 Brain Injury Criterion (BrIC)

BrIC is a criterion proposed by Takhounts et al. [45] [46]. The first version, referred to the abbreviation of BrIC, includes a combination of peak resultant rotational acceleration and peak resultant rotational velocity [45]. In 2013 [46], the criterion was updated, and the abbreviation for the updated criterion is BrIC. The updated criterion includes a combination of peak rotational velocity of the three main components (x,y,z) Formula (3). The critical values ($\omega_{cr,x}$, $\omega_{cr,y}$, $\omega_{cr,z}$) were defined based on computer simulations using two different FE models [SIMon (Simulated Injury Monitor) and GHBMC (Global Human Body Model Consortium)]. The pulses that were applied to the models were pulses from vehicle crashes (NCAP tests and frontal offsets tests). There has also been alternative critical values proposed of different research groups that is dependent on other FE models [47] [48] [49].

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{cr,x}}\right)^2 + \left(\frac{\omega_y}{\omega_{cr,y}}\right)^2 + \left(\frac{\omega_z}{\omega_{cr,z}}\right)^2} \quad (3)$$

Several studies have developed risk functions for BrIC. Takhounts et al. [46] developed risk functions for AIS1 to AIS5 based on scaled experiments of animal experiments. The risk was dependent on the critical values used. When the critical values where based on the cumulative strain density measures (CSDM), the 50 % risk for an AIS2 injury was 0,78 and an AIS4 injury 1,04. Wu et al. [50] also used animal experiments together with human volunteers and dummies to develop risk functions for mild and severe Traumatic Brain Injury (TBI). They found a 50 % risk for mild TBI (mTBI) of 0,77 and 1,19 for severe TBI (sTBI).

6.5 DAMAGE

Gabler at al. [48] developed DAMAGE (Diffuse Axonal, Multi-Axis, General Evaluation) to predict diffuse brain injuries with a second-order system of equations. The model use a mass-spring-damper system to predict the brain tissue strain due to rotational motion around the three principal axes of the head. The parameters in the second-order system of equations were determined by uniaxial and multiaxial loadings of a FE model, namely the Global Human Body Model Consortium (GHBMC). The advantages of this metric compared to many other kinematics-based metrics are that it considers the kinematics over time and includes all three principal axes of the head. The metric can be seen as a simplified FE model, which is less computationally expensive than a regular FE model.

Wu et al. [50] developed risk functions for DAMAGE for mild and severe traumatic brain injuries based on animal, human and dummy experiments. They found a 50 % risk of mTBI to be 0,359 for mTBI and 0,617 for severe TBI.

6.6 Tangential force at the helmet surface

An indirect way to assess rotational motion transferred to the head during an oblique impact is to measure the normal and tangential forces acting on the oblique anvil and compute the friction between helmet shell and anvil. This value of friction is then used as injury risk estimate. This method is used in the motorcycle helmet rating method SHARP [51]. The method has been criticised to simplify the helmet to ground interaction and that the correlation between tangential force and rotational acceleration is low [52]. The reason to use the indirect method was to avoid the necessity of instrumentation in the headform capable of measuring rotational acceleration. Today, instrumentations that measure either

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