Workshop on R22 Motorcycle Helmet Standard

Injury Criteria for complex loading

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DECK Caroline, BOURDET Nicolas, WILLINGER Rémy
deck@unistra.fr
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HEAD INJURY CRITERIA IN TERMS OF
GLOBAL KINEMATIC PARAMETERS:

TRANSLATION
**GLOBAL PARAMETERS (TRANSLATION)**

1. **Early, Head and Brain Injury Risk Assessments Were Based on Translational Head Kinematics.**

2. **Maximum Resultant Linear Acceleration Was Studied as an Injury Predictor Variable Since It Was Found to Correlate with Skull Fracture.** *(Gurdjian et al. 1966, Lissner et al., 1960)*

3. **The Threshold for $a_{\text{max}}$ Depends on Its Application.** Maximum Linear Acceleration Is Used for Many Years and Continues to Be Used in Several Helmet Standards *(CEN, Snell 1995, CSA 1985)*

4. **The Severity Index** *(Gadd 1966)*

$$\int a^{2.5} dt < 1200 \quad \text{NOCSAE STANDARD}$$
A variation of this criterion is A3MS value that refers to the maximum deceleration that lasts for 3ms:

- A3MS should not exceed 80G (Got et al., 1978).
- According to Chin et al. (1998), and based on COST 327 reports, a head acceleration of 200 to 250G and 250 to 300G leads respectively to severe AIS4, respectively AIS5 head injury;

The previous researches led to the development of the Wayne State Tolerance Curve (WSTC) (1966)

\[
HIC = \max_{(t_1, t_2)} \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right\}^{2.5}
\]

Head mass = 4.58 kg  
HIC = 1000
**Motorcyclist (2002) ECE R022**
Headform; \( V = 7.5 \text{ m/s} \);
- HIC < 2400
- PLA ≤ 275G

**Motorcyclist DOT FMVSS 218**
Headform; \( V = 6 \text{ m/s} \);
- PLA < 400G
- Dwell time for an acceleration of 200G must not exceed 2 ms;
- Dwell time for an acceleration of 150G must not exceed 4 ms
LIMITATIONS OF EXISTING MOTORCYCLE STANDARD IN TERMS OF HEAD INJURY CRITERIA FOR SHOCK ABSORPTION

- Based on translation parameters only
- No consideration of rotational acceleration
- Not direction dependent
- Not injury mechanism related
- Poor correlation with real world observation

HIC
PLA < 275g

\[ a_{350g} \leq 4\text{ms} \]
\[ a_{200g} \leq 2\text{ms} \]
HEAD INJURY CRITERIA IN TERMS OF GLOBAL KINEMATIC PARAMETERS:

ROTATION
It is **well known** that brain is sensitive to rotational acceleration since Holbourn (1943).

This phenomenon has essentially been addressed qualitatively with **animal** or physical **models**.

<table>
<thead>
<tr>
<th>Study</th>
<th>Species/Condition</th>
<th>Acceleration Range</th>
<th>Angular Velocity Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gennarelli, Thibault, Ommaya</td>
<td>25 Monkeys alive</td>
<td>1800 rad/s² to 7500 rad/s²</td>
<td>60 rad/s to 70 rad/s</td>
</tr>
<tr>
<td>Pincemaille et al. (1989)</td>
<td>Boxers training</td>
<td>13600 rad/s² to 16000 rad/s²</td>
<td>28 rad/s to 48 rad/s</td>
</tr>
<tr>
<td>Gennarelli et al. (1982)</td>
<td>More than 100 primates alive</td>
<td>15000 rad/s²</td>
<td>150 rad/s</td>
</tr>
<tr>
<td>Margulies et al. (1989)</td>
<td>Based on Gennarelli et al. (1982)</td>
<td>16000 rad/s²</td>
<td>46.5 rad/s</td>
</tr>
</tbody>
</table>

Holbourn (1943)
Global Parameters (Rotation)

**BrIC** (*Takhounts et al., 2011*):

\[
BrIC = \frac{\omega_{\text{max}}}{\omega_{cr}} + \frac{\alpha_{\text{max}}}{\alpha_{cr}}
\]

**BrIC** (*Takhounts et al., 2013*):

\[
BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xc}}\right)^2 + \left(\frac{\omega_y}{\omega_{yc}}\right)^2 + \left(\frac{\omega_z}{\omega_{zc}}\right)^2}
\]

**RIC Rotational Injury Criterion** (*Kimpara et al., 2011*):

\[
RIC = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \alpha(t) dt \right\}^{2.5} \right]_{\text{max}}
\]

- **Based on scaled animal data**
- **Only rotational contribution**
- **No direction dependant**
- **Based on a simple FE head model**
- **Dummy database to develop BrIC injury curves for each dummy...**
- **Based on 67 scaled animal data**
- **Same limitations than for SIMon**
- **BrIC was established for AIS4+**
- **Scaling of risk curves (HIC)**
- **Problem with critical values**
- **...**
HEAD INJURY CRITERIA IN TERMS OF GLOBAL KINEMATIC PARAMETERS:

COMBINED PARAMETERS
**Combined Parameters**

**GAMBIT** *(Newman et al., 1986)*:

\[ G(t) = \left[ \left( \frac{a(t)}{250} \right)^n + \left( \frac{\alpha(t)}{25000} \right)^m \right]^{\frac{1}{s}} \]

- \( n = m = s = 2.5 \)
- \( G=1 \) is set to correspond to a 50% probability of MAIS 3
- GAMBIT was never extensively validated as an injury criterion

**HIP** *(Newman et al., 2000)*: **Head Impact Power**

\( HIP = ma_x \int a_x dt + ma_y \int a_y dt + ma_z \int a_z dt + I_{xx} \alpha_x \int \alpha_x dt + I_{yy} \alpha_y \int \alpha_y dt + I_{zz} \alpha_z \int \alpha_z dt \)

- A 50% probability of concussion for \( HIP_{max} = 12.8 \text{ kW} \)
- \( HIP_{max} \) is not validated for more severe brain injuries

**PRHIC** *(Kimpara et al., 2011)*: **Power rotational head injury criterion**

\[ PRHIC = \left( t_2 - t_1 \right) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} HIP_{ang}(t) dt \right\}^{2.5} \]

- \( PRHIC_{36} \) computed from 31 impact events involving 58 American football players
- 50% risk of mild TBI value is \( PRHIC_{36} = 8.70 \times 10^5 \)
**COMBINED PARAMETERS**

**KLC : Kleiven’s Linear Combination** *(Kleiven et al., 2007)*:

\[
KLC = \frac{4718}{10^6} \omega_m + \frac{224}{10^6} HIC_{36}
\]

\( m \) is the maximum resultant rotational velocity

*Based on 58 NFL cases analysis*

**PCS : Principal Component Score** *(Greenwald et al., 2008)*:

\[
PCS = 10 \cdot ((0.4718 \cdot sGSI + 0.4742 \cdot sHIC + 0.4336 \cdot sLIN + 0.2164 \cdot sROT) + 2)
\]

PCS is a weighted sum of translation and rotational accelerations, HIC, and SI with empirically determined weights

*Based on NFL cases analysis*

**CP : Combined Probability of Concussion** *(Rowson et al., 2013)*:

\[
CP = \frac{1}{1 + e^{-\left(-10.2 + \frac{433}{10^4} a + \frac{873}{10^8} \alpha - \frac{92}{10^8} a\alpha\right)}}
\]

\( a \) is peak linear acceleration
\( \alpha \) is peak rotational acceleration

*Based on HITS data and 58 NFL cases analysis*
GLOBAL PARAMETERS-ROTATION

U-BRiC
Gabler et al. 2018

- Rotation only,
- Maximum value
- Critical value of Max
- Time evolution...\( \Delta t \)
- Natural period of brain (40ms)
- Based on a given brain model

![Image](image.jpg)

**MATERIALS AND METHODS**

**Development of U/BiC**

U/BiC is based on the assumption that maximum brain deformation under rotational head motion is analogous to deformation from a second-order system under excitation. In a previous study, a sDOF model was used to show that brain deformation in one dimension is governed by three general categories of rotational head motion, each distinguished by its pulse duration (\( \Delta t \)) relative to the natural period (\( \Delta t_n \)) of the brain–skull system: for short-duration pulses, maximum brain strain depended primarily on the magnitude of angular velocity (Fig. 1a, \( \Delta t \rightarrow \Delta t_n \)), for long-duration pulses, maximum brain strain depended primarily on the magnitude of angular acceleration (Fig. 1a, \( \Delta t \rightarrow \Delta t_n \)), and for pulses near the natural period of the brain (36–45 ms), maximum strain depended on the magnitudes of velocity and acceleration (Fig. 1a, \( \Delta t \rightarrow \Delta t_n \)).

To generalize the transition between velocity and acceleration dependent brain strains for a one-dimensional impact pulse, exponential functions were used (Fig. 1b). Adding these exponentials resulted in a function that switches between velocity and acceleration dependent deformations in a manner creating a velocity-only dependence in short-duration (\( \Delta t \rightarrow 0 \)) and acceleration-only dependence in long-duration, (\( \Delta t \rightarrow \infty \)).

\[
f(\Delta t) = a_1 (1 - e^{-\Delta t}) + a_2 e^{-\Delta t},
\]

where \( f \) is a functional that establishes brain deformation given the magnitudes of angular velocity (\( \omega \)) and angular acceleration (\( \alpha \), and duration of a one-dimensional head impact pulse. Assuming that the development of the metric is related to the magnitudes of angular velocity and acceleration, \( \Delta \omega = \omega_0 \), and that the one-dimensional deformation can be generalized for rotations about each axis of the head, the following kinematic-based metric is proposed:

\[
U/BiC = \left\{ \sum \left[ \omega_i^* \left( \frac{\alpha_i^* - \omega_i^*}{\omega_i^*} \right) \right] \right\}^{1/2},
\]

where \( \omega_i^* \) and \( \alpha_i^* \) are the directionally dependent (\( \gamma = x, y, z \)) maximum magnitudes of head angular velocity and angular acceleration, each normalized by a critical value (\( \omega_0 \), \( \alpha_0 \)). The critical values normalize the metric to maximum brain strain from a FE model and control the transition between velocity and acceleration dependent deformations. The exponent \( r \) establishes the power at which the magnitude is evaluated; model performance was assessed for \( r \) equals one and two. Six total parameters (two critical values per direction) were used to establish the full three-dimensional form of U/BiC.

**Development of a Metric for Probing Brain Strain**

**Additional Kinematic Forms**

In addition to U/BiC, several mathematical forms based on existing rotational metrics were assessed for predicting brain strain responses. These metrics are based on the maximum (resilient or directionally dependent) magnitudes of angular velocity and angular acceleration, and were included in the analysis to benchmark improvement using U/BiC with data from the current study (Table 1). Metrics based on translational kinematic parameters were not included, since they were shown to have poor correlation with strain-based metrics.36
GLOBAL PARAMETERS-ROTATION

U-BRiC
Gabler et al. 2018

Gabler et al.

(a) Correlations with database

<table>
<thead>
<tr>
<th>Parameter</th>
<th>MPS</th>
<th>CSDM</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\omega_m^*$</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$\alpha_m^*$</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$C_m$</td>
<td></td>
<td></td>
</tr>
<tr>
<td>BrIC (refit) (peak)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>BrIC (refit) (p2p)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$\alpha_i^*$</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$C_i$ (peak)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$C_i$ (p2p)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>UBrIC (peak)</td>
<td></td>
<td></td>
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<tr>
<td>UBrIC (p2p)</td>
<td></td>
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</tr>
</tbody>
</table>
GLOBAL PARAMETERS - ROTATION

U-BRiC
Gabler et al. 2018

FIGURE 3. Scatter plots for top performing fitted metrics to MPS (left column) and CSDM (right column) using the database of 1595 head impacts. MPS is based on the 95th percentile value. Solid red lines indicate a one-to-one relationship, while dotted red lines are ±1 root mean square error. Results shown are for metrics with r = 2.
### Conclusions on Global Parameters

- **A Number of Studies** focused on the victim kinematics in real-world accident and demonstrated the effectiveness of tangential head impact conditions (*Mills et al.* (1996), *Bourdet et al.* (2011, 2012)…);

- **Despite this Consolidated Knowledge** no head protection standard are currently considering head rotational acceleration;

- **There is No Relevant** combined, time, and direction dependent brain injury criteria in terms of global head acceleration;

- **A Number of Tentative Exist**: Based mainly on two types of database

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Brain Injury Criteria</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\alpha_{max}$</td>
<td>(Gennarelli et al., 1972, 1982)</td>
</tr>
<tr>
<td>BrIC</td>
<td>(Kahnemann et al., 2011, 2013)</td>
</tr>
<tr>
<td>HIC</td>
<td></td>
</tr>
</tbody>
</table>

- **There is a Need to Set Properly**:
  - A tissue level brain injury criteria
  - A measure of the quality of an injury criteria
FINITE ELEMENT HEAD MODELS:

HUMAN HEAD IS NOT JUST A SENSOR

FINITE ELEMENT MODELING
I NJURY METRICS BASED ON ANIMAL DATABASE

WSU FEM
King et al., 2003

SIMon FEM
Tackounts et al., 2007

KTH FEM
Kleiven et al., 2007

THUMS FEM
Atsumi et al., 2016

INJURY METRICS BASED ON NFL DATABASE

INJURY METRICS BASED ON REAL WORLD DATABASE

Dublin FEM
Gilchrist et al., 2004

KTH FEM
Kleiven et al., 2007

THUMS FEM
Atsumi et al., 2016

GHBMC
Combest et al., 2016

SUFEHM
Sahoo et al., 2016

DHIM
Zhao et al., 2018

BrIC

DAMAGE

University of Strasbourg

Injury Criteria for complex loading
DATABASES USED TO DERIVE HEAD INJURY CRITERIA

RIC (Kimpara et al., 2011)
CP (Rowson et al., 2013)
PCS (Greenwald et al., 2008)
KLC (Kleiven et al., 2007)
PRHIC (Kimpara et al., 2011)
HIP (Newman et al., 2000)
GAMBIT (Newman et al., 1986)

WSU FEM (King et al., 2003)
KTH FEM (Kleiven et al., 2007)

Strain * Strain rate
First Principal Strain
Axonal strain

Animal database

$\alpha_{max}$
BrIC (Takhounts et al., 2011, 2013)
HIC
$\omega_{max}$ (Gennarelli et al., 1972, 1982)

SIMon FEM (Takhounts et al., 2007)

CSDM, RMDM

NFL database

Dummy Responses

DAMAGE (Gabler et al., 2011)

Real World Data

SUFEHM (Sahoo et al., 2016)
Axonal strain
Von Mises stress
FOUR TYPES OF DATABASES ARE USED TO DERIVE HEAD INJURY CRITERIA FOR GLOBAL PARAMETER METRICS AS WELL AS FEM PARAMETERS:

- ANIMAL DATABASE
- DUMMY DATABASE
- NFL DATABASE
- REAL WORLD ACCIDENTS DATABASE

FOR EACH IT EXISTS SOME LIMITATIONS:

- HUMAN = ANIMAL?
- NOT INJURY RELATED
- REPEATED IMPACTS, RECORD SYSTEM, CONCUSSION (LOW SEVERITY)

MOST CONFIDENT DATABASE IS A REAL WORLD ACCIDENT DATABASE WITH DIFFERENT LOADINGS, SEVERITY, INVOLVING HUMAN…

ONLY ONE FEM HAS HEAD INJURY CRITERIA BASED ON REAL OBSERVATIONS: SUFEHM
SUFEHM (Strasbourg University Finite Element Head Model)

- Brain Cerebellum (5320 brick elements)
- Brainstem (188 brick elements)
- Skull (1797 shell elements)
- Membranes (471 shell elements)
- Cerebrospinal fluid (2591 brick elements)
- Scalp (2294 brick elements)
- Face (530 shell elements)
125 real world accidents

Analitical or experimental replication

Numerical simulation

Derivation of tolerance limits to specific injury mechanisms
125 accidents, 2 injury mechanisms (mild brain injury or loss of consciousness and skull fracture), approx. 20 parameters calculated per head trauma.

Binary logistic regression
Nagelkerke R-square statistics (R²)
TOLERANCE LIMIT FOR SKULL FRACTURE

Risque de fracture crânienne (%)

Energie de déformation du crâne (mJ)

Skull internal energy (mJ)

Robustness

$R^2 = 0.633$

448 MJ

Injury Criteria for complex loading
Robustness $R^2=0.876$

✓ TOLERANCE LIMITS IN TERMS OF AXONAL STRAIN ABOUT 14.65% FOR 50% RISK OF AIS2+ (IN ACCORDANCE WITH EXPERIMENTAL VALUES)

✓ WITH HOMOGENEOUS BRAIN MECHANICAL PROPERTIES A TOLERANCE LIMIT IN TERMS OF VON MISES STRESS ABOUT 37kPA FOR 50% RISK OF AIS2+
VERY POOR CORRELATION OF HIC!

FOR SKULL FRACTURE

FOR MTBI (AIS2+)

ROBUSTNESS OF TISSUE LEVEL CRITERION VERSUS HIC
Conclusions
A number of **limitations** exist on metrics based on **global parameters**.

**Need to go** **further** than **HIC calculation** for **R 22 06**.

**Need to take into account:**
- 3D linear components **and**
- 3D angular components **and**
- Impact location/direction
- Tissue level injury criterion (bone, brain…)

**Finite element models are powerful:**
- **Model specifications** can be proposed
- Defined **robustness of criteria**
- **Type of database** to be used

**Recommendation proposal:** To **monitor SUFEHM** in the future standard based on **6D headform loadings vs time**.
How to easily use a predictive head FEM

Sufehm Toolbox

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