Correlation of Global Head and Brain Tissue Injury Criteria to Experimental Concussion derived from Monkey Head Trauma Experiments

Jacobo Antona-Makoshi , Johan Davidsson, Susumu Ejima, Koshiro Ono, Karin Brolin, Kenji Anata

Abstract A series of 24 frontal head traumatic impacts on macaques carried out in the past were simulated with a validated finite element model of the specimens. From these simulations, brain tissue response and head accelerations were extracted. Based on the accelerations, global head injury criteria were calculated. Correlation between the brain tissue mechanical parameters, the global head injury criteria and the concussion scored in the experiments were analyzed. Based on this analysis, global head injury criteria that best correlate with concussion score for frontal impacts were identified and injury risk functions for brain tissue that can be used for human FE models are proposed. In addition, the new results were compared to a previous study based on simulations of 19 occipital head impacts from the same data source.

Keywords Concussion, finite element model, head, monkey experiments, traumatic brain injury.

I. INTRODUCTION

Fatalities and severe injuries due to road traffic accidents are still a serious health and economic issue in today's society, where brain injuries are one of the most common severe injuries. However, improved accident avoidance systems are predicted to mitigate and reduce accident severity, thus giving more focus to long-term disabling injuries of which Traumatic Brain Injury (TBI) is a major issue. To develop effective countermeasures in head impacts, it is essential to understand TBI mechanisms and establish associated thresholds. In the past, head impact experiments on non-human primates (NHP), used as human surrogates, were carried out [1-4]. Some of these results were used in the development of the Head Injury Criterion (HIC) [5-6] currently in use in the FMVSS 208 regulation. The HIC has been studied and utilized for years, but is still criticized for not considering all factors that are important to brain injury. Such additional factors include the impact direction and area of contact, stiffness of the impacting surface, and rotational accelerations induced by oblique impacts or when the torso is restrained [7]. Therefore, alternative or complementary criteria have been proposed that consider rotational acceleration of the head, such as the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) [8], the Brain Rotational Injury Criterion (BRIC) [9] and the Rotational Injury Criterion (RIC) [10]. The two latter were proposed, together with brain tissue injury criteria, for dummies and human head Finite Element (FE) models and are undergoing validation with reconstructions of real-life sports and traffic accidents, and scaled animal injury data. Unfortunately, the accuracy of the methods used to collect information on head kinematics and precise injury severity and location from real-life events has limitations [9][11] and complementary forms of validation are needed to provide trustful criteria and associated thresholds.

By re-analyzing and reproducing existing NHP head experiments, using a model of these specimens, the reliability of suggested injury criteria can be evaluated. This approach was adopted in a previous study [12] to simulate and analyze a sub-set of 19 occipital NHP head impacts selected from 149 trauma experiments conducted at the Japan Automobile Research Institute (JARI) in the past [4] that also included frontal [4][13] and lateral [14][15] head impacts. The Antona et al. [12] study identified brain tissue mechanical parameters, such as the Von Mises Stress (VMS) and the Maximum Principal Strain (MPS), at the cerebrum and the brain stem to have a correlation to occurrence of concussion. In addition, brain tissue injury risk functions for concussion were proposed. Based on the assumption that tissue thresholds are the same for NHP and humans,

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such injury risk functions can be used to interpret results obtained with human FE models.

However, several issues in [12] remained unresolved. First, the number of simulated cases was too low to draw sufficiently confident corridors for the brain tissue injury risk functions. Second, all simulated impacts were occipital and it was uncertain if the results could be generalized to other types of impacts, such as frontal or lateral. Third, rotational acceleration curves from the reconstructed occipital experiments were missing; therefore an analysis of global head injury criteria and their relation to experimental injuries was not conducted.

Therefore, to resolve these issues, the same simulation-based approach can be applied to other sub-sets of the JARI NHP head trauma experiments. In addition, head linear and rotational acceleration curves can be extracted from the simulations, verified against re-processed experimental high-speed films, and used to calculate the global head injury criteria: HIC, BRIC, RIC and GAMBIT. These criteria, in contrast to tissue criteria, can be used with crash test dummies. Assuming that similarity principles between NHP and humans apply for the global head kinematics, the calculated injury criteria combined with the injury data from the original NHP experiments can provide an additional evaluation of these global head injury criteria.

Based on this background, the aims of this work are:

1. To provide brain tissue injury risk values for concussion, that can be applied as reference for injury predictions in the analysis of frontal impacts, with human FE models.
2. To evaluate how the existing global head injury criteria perform compared to experimental concussion type injury in frontal impacts with NHP.

II. MATERIALS

In this study, an FE model of a NHP head and neck that was previously developed and validated by [12] was used to reconstruct a sub-set of 24 frontal NHP head impacts from the JARI head trauma experimental database. All simulations were performed with the explicit FE code LS-DYNA [16]. Post-processing and analysis were done with the following codes: LS-Prepost (LSTC, Livermore, US), Microsoft Excel (Microsoft Corporation, US) and R-statistic (The R Foundation for Statistical Computing, Vienna, Austria) [17].

Monkey FE Model

The Monkey FE model in Figure 1 is described in detail by [12]. In short, the geometry was developed based on medical images, and material models and material properties were implemented according to Table I. The model was validated at the tissue level with experimental compression data from coupons of monkey scalp [18] and brain tissue [19][20], at the component level with quasi-static head compression test data by [21], and for head kinematics with full-scale head impact experiments by [4].

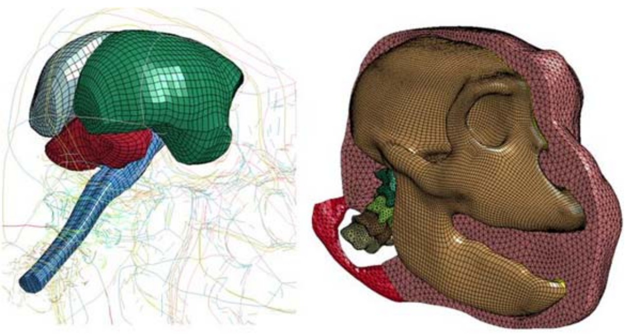


Fig. 1 The Monkey FE model. Images of the brain (left) and the skull, cervical spine and soft tissues (right) where part of the flesh was removed for visibility reasons.

TABLE I
MONKEY FE MODEL MATERIAL PROPERTIES

	Material model	Material properties
Scalp and neck flesh	Fu Chang Foam	Experimental stress-strain curves
Skull bone	Piecewise linear plasticity	E = 6.48 GPa, US = 92.4
Cerebrum, corpus callosum, cerebellum	General viscoelastic	G0 = 10300 Pa, G1 = 3700 Pa, tau= 100 s-1
Brain stem	General viscoelastic	G0 = 18540 Pa, G1 = 6660 Pa, tau= 100 s-1

E=Young Modulus, US=Ultimate strength, G0=Short term modulus, G1=Long term modulus, tau=decay constant

Sub-group of 24 frontal head trauma experiments

A consistent sub-set of data was selected from the JARI head trauma experimental database using the following criteria:

- Only direct frontal impacts,
- The specimens were either *Macaca Fuscata* (Japanese monkey) or *Macaca Mulatta* (Rhesus monkey),
- The specimen had not sustained skull fracture, and
- Impact conditions (impact velocity, impactor type and maximum stroke) were available.

A group of 24 impacts fulfilled the selection criteria, see Table III (first four columns). For this group, specimen's anatomical data, impact conditions, head accelerations, high speed videos, photographs from pathological examinations and injury reports were available. The impacts were delivered to the specimen's forehead in a direction almost parallel to the specimen's Frankfort line by a 13 kg impactor. Pre-defined impact speeds ranged from 8.4 to 28.7 m/s. The head of the impactor was a rubber block, with stiffness ranging from intermediate (C) to soft (E) according to the material properties in [4]. A stopper was used to set the variable maximum impact strokes, ranging from 20 to 90 mm. The impacts resulted in combined translational and rotational head accelerations. Linear acceleration curves were captured and reported for 20 cases, but rotational acceleration curves were missing. The high speed films were captured with a 16 mm high-speed camera (HYCAM, Redlake; STALEX, Weinberger) at 4,000 frames per second.

The reported injuries included subdural hematoma, subarachnoid hematoma, contusions and concussions. The severity of a concussion was evaluated by measuring the physiological changes right after impact, according to a definition in use at the time of the experiments [22]. To quantify concussion, the following three criteria were used:

- Persistent loss of corneal reflex for at least 20 seconds after the impact,
- Cessation of respiration for at least 20 seconds after the impact, and
- Two levels of blood pressure disturbance following the impact.

In the original study, concussion grade was assigned when none of the three criteria applied. Concussion grades I, II, or III were assigned when one, two or three criteria applied. In the work presented in this article, all the cases that presented concussion were grouped together. According to this method, 10 out of the 24 impacts reported concussion injury.

III. METHODS

The overall methodology of this work is illustrated in Figure 2 and further details are provided below. The methodology consists of four steps:

- Case-by-case simulation of 24 frontal impact experiments,
- Comparison of head motion in the experiments and the simulations,
- Calculation of global head injury criteria based on the simulations, and
- Statistical analysis of the results to establish the correlation between:
  - Global head injury criteria and experimental concussion (number 1 in Figure 2),
  - Brain tissue injury criteria and experimental concussion (number 2 in Figure 2), and
  - Global head injury criteria and brain tissue injury criteria (number 3 in Figure 2).

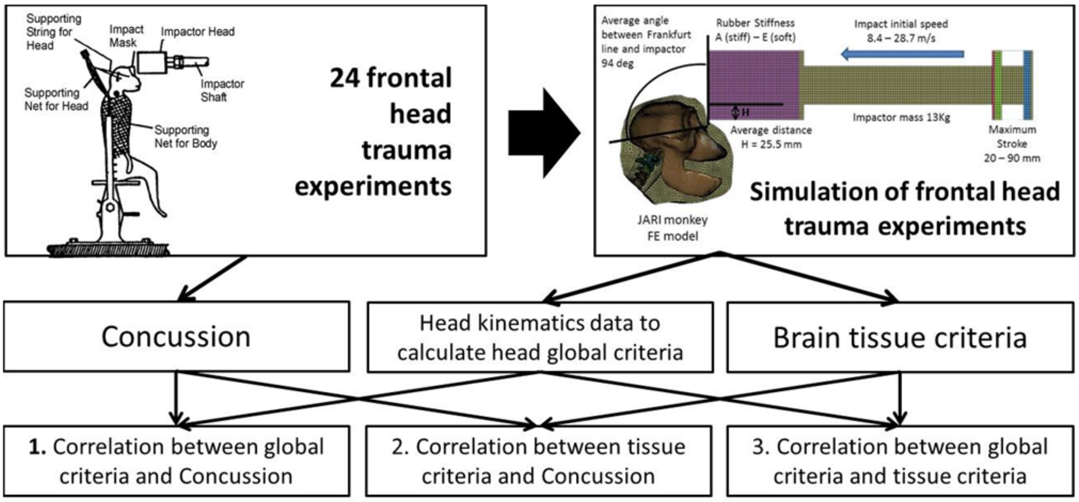


Fig. 2 Scheme of the methodology used in this study, illustrating the three statistical analyses.

Case-by-case simulation of 24 frontal impact experiments

Each of the 24 experimental cases was simulated with the monkey FE model in the setup illustrated in Figure 3. The angle and vertical distance that defined the initial position of the monkey FE model with respect to the impactor were kept constant at 94 degrees and 25.5 mm according to the measurements reported in the original experiments. The velocity of the impactor just before the impact, the stiffness of the rubber block, and the maximum stroke of the impactor were set case-by-case according to the reported data from the experiments [4] and listed in Table III (columns 1-4). All the simulations were run for the entire time period of contact between the head and the impactor and for at least 2 ms more. Further details on the techniques used to implement the experimental boundary conditions can be found in [12]. Finally, the simulation results were processed to extract head translational and rotational accelerations, and peak values for VMS and MPS at the cerebrum and the brain stem.

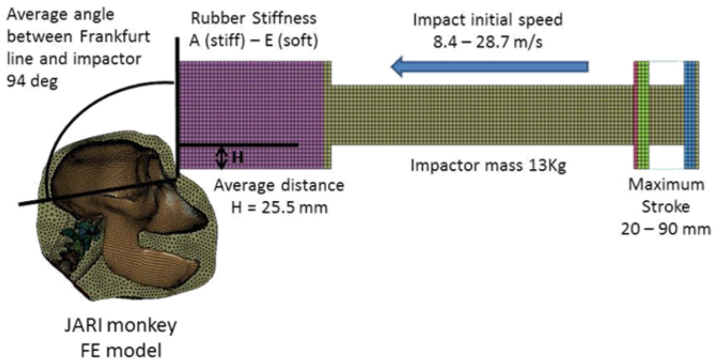


Fig. 3 Scheme of the simulation setup defined based on the reported data from the tests.

Comparison of head motion in the experiments and the simulations

The motions in the saggital plane of the heads in the experiments were retrieved through film analysis of the original experimental high-speed films (MOVIAS Neo Ver.2.10, NAC Image Technology Inc., Japan) and compared to the numerical resulting from the case-by-case simulations. The displacement of the head center of gravity and head rotation were output.

Calculation of global head injury criteria based on the simulations

The acceleration curves obtained in the case-by-case simulations were used to calculate HIC, GAMBIT, BRIC and RIC for each of the 24 frontal impacts. Table II presents the head injury criteria, their equations and a description of the parameters required to calculate each criterion. Acceleration thresholds at which injury occurs in NHP are higher than for humans and, therefore, the parameters used to calculate GAMBIT and BRIC were adapted to known values for macaques reported by [23] and [24].

TABLE II

GLOBAL HEAD INJURY CRITERIA

Criterion	Equation	Parameters
HIC [5][6]	$HIC = \left\{ \left[\frac{1}{t2 - t1} \int_{t1}^{t2} a(t)dt \right]^{2.5} (t2 - t1) \right\}_{max}$	a(t) from simulated impacts
GAMBIT [8]	$GAMBIT = \left[\left(\frac{a_{max}}{a_{cr}} \right)^2 + \left(\frac{\alpha_{max}}{\alpha_{cr}} \right)^2 \right]^{\frac{1}{2}}$	a_{max} and α_{max} from simulated impacts; $a_{cr} = 350g$ from [24] and α_{cr} = 12000 rad/s2 from [23]
BRIC [9]	$BRIC = \frac{\omega_{max}}{\omega_{cr}} + \frac{\alpha_{max}}{\alpha_{cr}}$	α_{max} and ω_{max} from simulated impacts. ω_{cr} = 140 rad/s and α_{cr}= 12000 rad/s2 from [23]
RIC [10]	$RIC = \left\{ \left[\frac{1}{t2 - t1} \int_{t1}^{t2} \alpha(t)dt \right]^{2.5} (t2 - t1) \right\}_{max}$	$\alpha(t)$ from simulated impacts

a(t): Linear acceleration; a_{max}: Maximum linear acceleration; a_{cr}: Critical linear acceleration ; $\alpha(t)$: Rotational acceleration; α_{max}: Maximum rotational acceleration; α_{cr}: Critical rotational acceleration; ω_{max}: Maximum rotational velocity; ω_{cr}: Critical rotational velocity

Statistical analysis

First, a T-test for significance was carried out to evaluate the relationship between the global head injury criteria and the experimental concussion. Second, brain tissue injury criteria were correlated with the experimental concussion and injury risk curves were developed (assuming log-logistic distribution) by applying survival analysis to account for censored data. This was done according to ISO/TC22/SC12/WG6 guidelines described in [25]. Third, linear regression between the global head injury criteria and the brain tissue injury criteria was conducted based on simulation results and the criteria were ranked by magnitude of correlation coefficient. The statistical analysis was carried out with the frontal impacts and compared to the results from occipital impacts [12]. Finally, the same analysis was carried out on the combined frontal and occipital impact data.

IV. RESULTS

Comparison of head motion in experiments and simulation

The impacts resulted in a head motion in the sagittal plane, combining a rear- and downward translation with an extension rotation. Figure 4 gives one representative example of the comparison between the head translation and rotation extracted from the films and the simulations for an intermediate severity impact (case 336 in Table III).

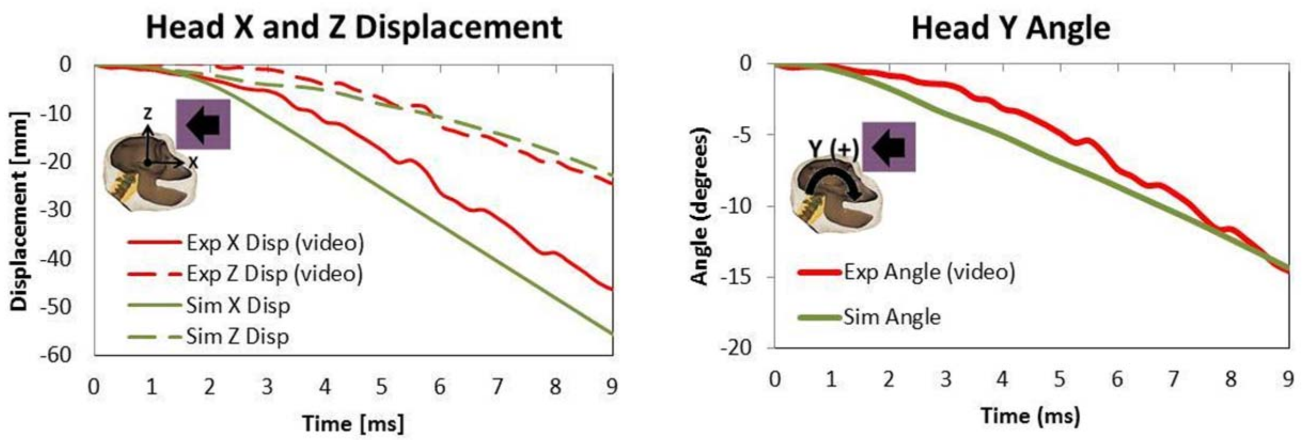


Fig. 4 Comparison of head linear (left) and angular (right) displacements from the simulations (in green) and the experiments (in red)

Case-by-case simulation of 24 frontal impact experiments

Table III lists the experiment number (column 1, consistent with experiment numbering from [4]) impact conditions (columns 2 to 4) and the injury measurements including physiological changes and scored concussion (columns 5 to 8) from the former experiments [4]. The table also includes the global head injury criteria (columns 9 to 12) and the brain tissue response (columns 13 to 16) from the simulations presented in this article.

TABLE III

FRONTAL IMPACT TEST CONDITIONS, INJURY OUTPUT AND SIMULATION RESULTS

	From experiments ([4])							From simulations							
	Impact conditions			Injury measurements				Global head injury criteria				Brain tissue parameters			
												Cerebrum		Brain stem	
Exp No	Impact velocity (m/s)	Rubber block (1)	Impact Stroke (mm)	Apnea (s)	Blink (s)	Brady cardia (2)	Concussion (Grade) (3)	HIC	GAMBIT	BRIC	RIC	VMS (Pa)	MPS	VMS (Pa)	MPS
305	10.5	C	20	5	22	No	Yes (I)	1818	1.4	1.1	2.5E+8	3150	0.19	3471	0.15
306	12.9	C	20	8	8	No	No (0)	2628	1.7	1.4	4.6E+8	3213	0.20	3713	0.16
307	15.5	C	20	5	16	No	No (0)	3720	2.2	1.8	7.8E+8	3364	0.21	4794	0.21
308	17.5	C	20	15	13	Yes	Yes (I)	4688	2.6	2.2	1.1E+9	3481	0.22	5630	0.24
309	18.8	C	20	3	10	No	No (0)	5374	2.9	2.5	1.3E+9	3555	0.22	6088	0.26
325	15.7	C	30	0	12	No	No (0)	7682	2.3	1.9	8.4E+8	3432	0.21	5391	0.23
326	20.0	C	30	30	30	Yes	Yes (III)	11546	3.3	2.8	1.7E+9	3719	0.22	7184	0.31
328	15.5	E	30	0	15	No	No (0)	2900	1.6	1.3	3.7E+8	3202	0.20	3372	0.15
329	19.6	E	30	2	20	No	Yes (I)	4453	2.2	1.9	7.6E+8	3375	0.21	4348	0.19
331	15.5	C	30	0	10	No	No (0)	7502	2.3	1.8	8.1E+8	3421	0.22	5299	0.23
332	19.0	C	30	0	16	No	No (0)	10589	3.0	2.5	1.5E+9	3639	0.22	6908	0.30
333	20.1	C	45	5	20	No	Yes (I)	21128	3.4	2.7	1.3E+9	5288	0.37	7042	0.31
336	19.4	E	30	8	15	No	No (0)	4374	2.2	1.9	7.3E+8	3363	0.22	4283	0.19
337	18.9	E	60	8	12	No	No (0)	10830	2.2	1.8	6.8E+8	3343	0.23	5656	0.24
338	18.4	E	90	10	10	No	No (0)	18298	3.0	2.7	1.3E+9	3669	0.26	5204	0.23
339	21.7	E	60	4	9	No	No (0)	12872	2.6	2.3	1.0E+9	3474	0.22	6399	0.28
341	24.7	E	60	0	20	No	Yes (I)	14917	3.1	2.8	1.5E+9	3686	0.26	6766	0.29
343	24.6	E	90	20	23	Yes	Yes (III)	20838	3.2	2.8	1.5E+9	4241	0.29	6926	0.30
365	19.9	D	60	10	10	-	No (0)	16591	2.7	2.2	9.8E+8	3915	0.29	6351	0.27
366	19.4	D	80	2	10	Yes	Yes (I)	24304	3.9	3.5	2.0E+9	4879	0.34	6331	0.27
367	25.3	D	40	5	11	No	No (0)	13414	3.6	3.2	1.9E+9	3766	0.27	7221	0.31
368	28.7	D	40	2	10	Yes	Yes (I)	15549	4.3	3.9	2.7E+9	3935	0.26	7959	0.34
373	8.4	E	50	6	12	No	No (0)	2347	1.0	0.8	1.0E+8	2993	0.19	2874	0.13
374	15.5	E	80	40	9	Yes	Yes (II)	12399	2.7	2.4	8.8E+8	3673	0.22	4452	0.19

(1) Impactor rubber block type according to [4].
(2) Bradycardia 'Yes' when two levels of blood pressure disturbance followed impact.
(3) Occurrence of concussion used in this study. In brackets, the concussion grade scored in the experiments [4]

Head and brain kinematics

Figure 5 illustrates the head kinematics and brain strains for one of the simulated injurious impacts (case 333 in Table III). In this simulated impact, a sudden increase of the linear acceleration occurred when the rubber block came in contact with the head. As the impactor advanced, the rubber block bent and deformed to adapt its shape to the forehead of the model. The time when this deformation is fully developed coincided with a plateau in the linear acceleration curve. Then, at about 2 ms, the impactor base was stopped as it reached the maximum pre-defined stroke. Due to the inertia of the rubber block and the energy stored in this block, its front surface continued to move and transfer energy to the head, causing a second rise in the linear head acceleration until a peak of 716 g was reached at 2.9 ms. After that, the head decelerated until the end of the impact at about 4.2 ms. As for the rotational acceleration, an early first peak occurred (29,900 rad/s^2 at 0.5 ms) right after the start of the impact. Then, it decreased during the linear acceleration plateau and rose again due to the rubber recoil effect, reaching a second peak acceleration (31,200 rad/s2 at 2.9 ms) that coincided in time with the peak in linear acceleration. During the impact, strains in the brain stem and the cerebrum increased gradually until they both peaked between 3.0 and 3.2 ms.

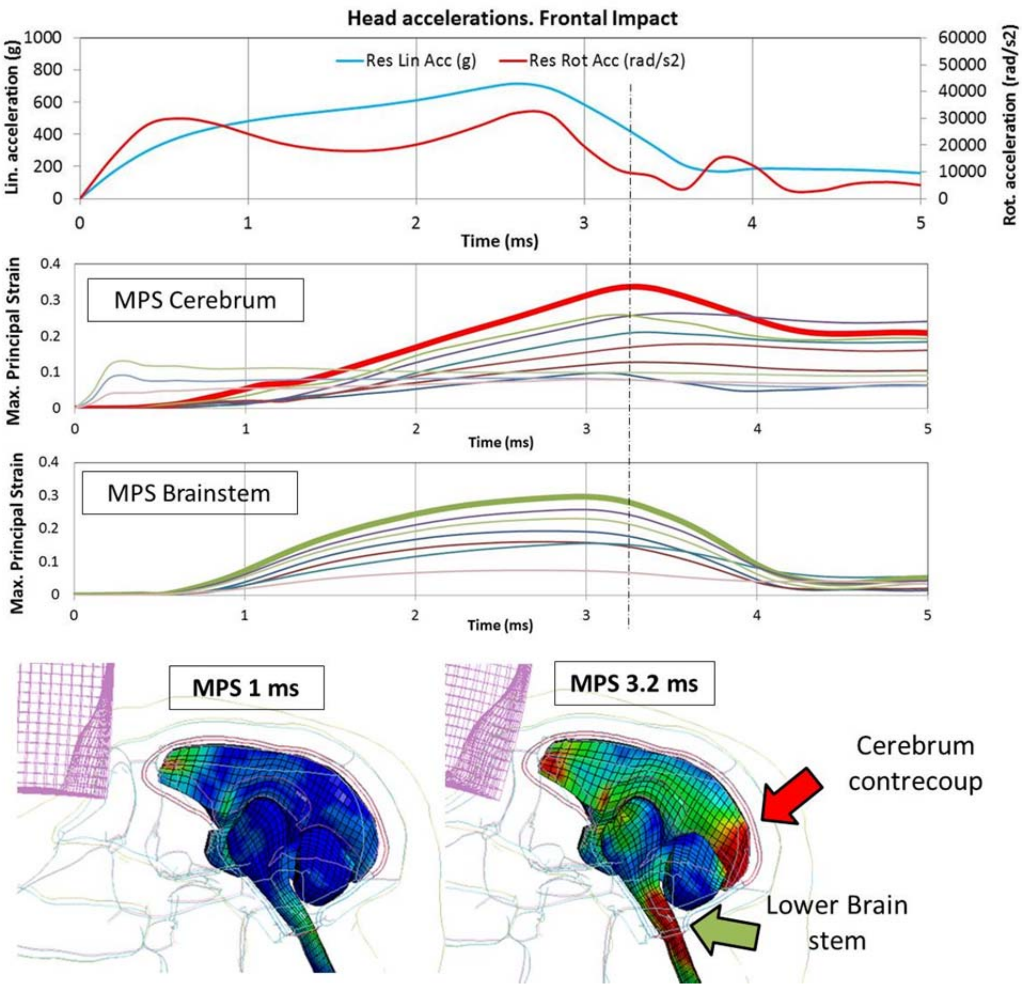


Fig. 5. Analysis of head kinematics in relation to brain tissue strains, case 333. Head linear and rotational acceleration during impact (top figure), maximum principal strain curves of several elements in the cerebrum and the brain stem (middle figures) including the one that showed the highest values (in red and green for the cerebrum and the brain stem, respectively) and fringe plots of the maximum principal strains (above 0.18 in red) in the brain at 1 ms (bottom left) and 3.2 ms (bottom right).

For later comparison with this study, the results from the occipital impacts in [12] are included in the appendix in a similar format to Table III and figure 5.

Global head injury criteria and experimental concussion

Figure 6 shows the mean values and standard deviations for each global head injury criterion, grouped according to the experimental injury outcome, for the frontal impacts simulated in this study (n=24). The results indicate that the averaged global head injury criteria were significantly higher for the group of cases that scored concussion (at a 95% confidence level).

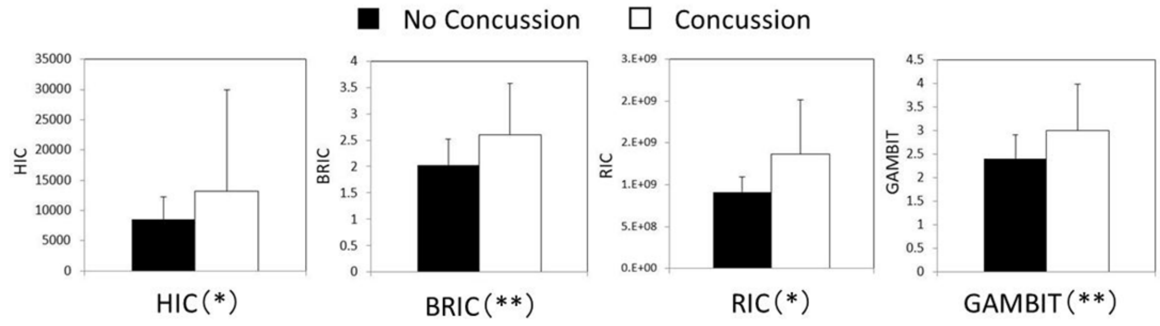


Fig. 6. Mean and standard deviation of the global head injury criteria grouped according to occurrence of concussion. Black bar: No Concussion. White bar: Concussion. (*) for p value < 0.05. (**) for p value < 0.99

Brain tissue injury criteria and experimental concussion

Figure 7 shows the injury risk curves and 95% confidence corridors for MPS at the brain stem for frontal head impacts, occipital head impacts and combined impacts obtained from the survival analysis. According to the curves deducted for frontal impacts, a 50% probability of concussion for frontal impacts corresponds with strain in the brain stem of 0.27. This value is higher than that obtained for the occipital impacts 0.19 (corrected from 0.21 in [12] by applying survival analysis instead of logistic regression). According to the quality index assessment based on relative size of the 95% confidence interval described in [25], the frontal data set provided *Unacceptable* confidence interval at 50% injury risk, while for occipital and combined impacts, *Fair* confidence intervals were obtained.

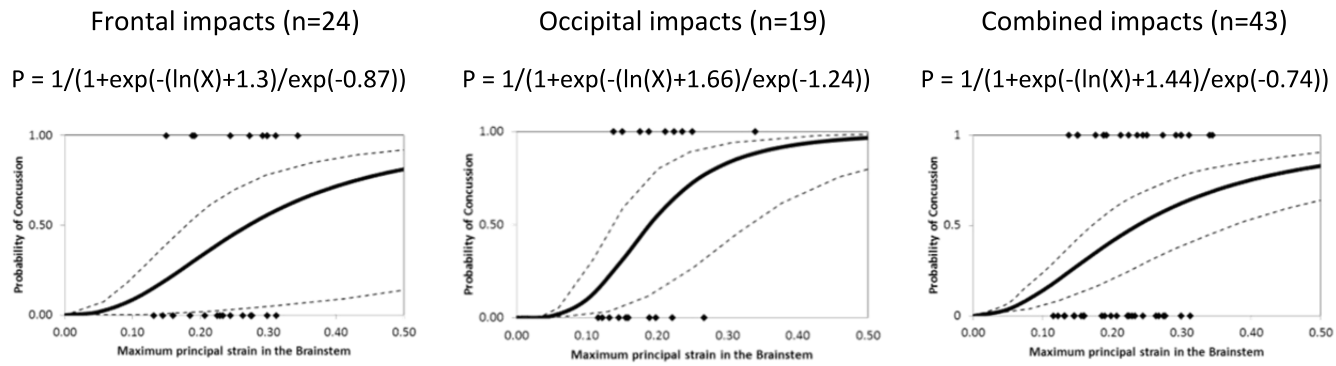
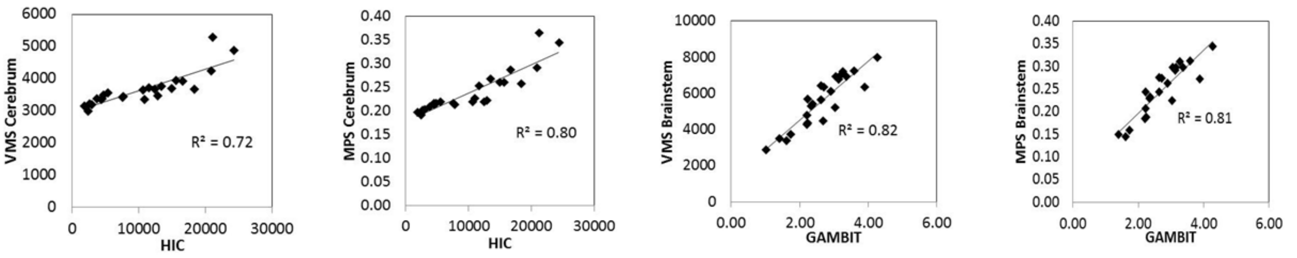


Fig. 7 Probability of concussion (solid line) and 95% confidence intervals (dotted lines) for maximum strain at the brain stem. Curves for frontal impacts (left colum) are compared to previous occipital impacts (middle column) and combined impacts together (right column).

Correlation between global head injury criteria and brain tissue injury criteria

Figure 8 shows the linear regression for the combinations of global head injury criteria and brain tissue response for the frontal impacts with the highest correlation coefficients (R^2). Table IV summarizes the correlation coefficients for the frontal impacts, the occipital impacts, and the combined frontal and occipital impacts. In the frontal impacts, HIC provided the highest correlation to the cerebrum strains while GAMBIT had the highest correlation for the brain stem strains. In the occipital impacts, GAMBIT provided the highest correlation for the brain stem strains while BRIC and GAMBIT were equally correlated to cerebrum strains. Combining the frontal and occipital impacts reduced the correlation coefficients of all indicators below the values obtained for either one or both of the frontal and occipital impacts.



VMS in the cerebrum vs HIC | MPS in the cerebrum vs HIC | VMS in the brain stem vs GAMBIT | MPS in the brain stem vs GAMBIT

Fig. 8. Linear regression between head kinematics and brain tissue criteria for frontal impacts with highest correlation coefficient

TABLE IV

CORRELATION COEFFICIENTS (R^2) BETWEEN HEAD KINEMATICS AND BRAIN TISSUE CRITERIA. (Bold print indicates the highest correlation for each column.)

	Frontal impacts (n=24)				Occipital impacts (n=19)				Combined impacts (n=43)			
	Cerebrum		Brain stem		Cerebrum		Brain stem		Cerebrum		Brain stem	
	VMS	MPS	VMS	MPS	VMS	MPS	VMS	MPS	VMS	MPS	VMS	MPS
HIC	**0.72**	**0.80**	0.52	0.51	0.82	0.82	0.66	0.66	**0.65**	**0.50**	0.44	0.48
BRIC	0.47	0.47	0.76	0.76	**0.94**	**0.94**	0.84	0.85	0.36	0.18	**0.82**	0.82
RIC	0.40	0.40	0.78	0.78	0.85	0.85	0.65	0.65	0.18	0.06	0.79	0.77
GAMBIT	0.53	0.51	**0.82**	**0.81**	**0.94**	**0.94**	**0.91**	**0.92**	0.51	0.30	0.81	**0.84**

V. DISCUSSION

In the frontal impact experiments reproduced in this study, the impactor hit the forehead of the specimens slightly above their head center of gravity. This affects the head accelerations; early rotational acceleration peaks appear to have been present in all frontal experiments (Figure 5). As the impact progressed, there were additional rotational acceleration peaks that were almost coincident in time with the linear acceleration peaks. In the simulations the highest brain tissue strains occurred in the occipital region of the cerebrum and in the rear side of the brain stem between the pons and the foramen magnum right after the linear accelerations peaked (Figure 5). It should be noted that, in the original work [4], contusions and sub- arachnoid hematoma in these two regions were reported. Hence, the performed frontal impact simulations are consistent with the hypothesis of brain stem damage being responsible for concussions as suggested by Kanda et al. [13]. The analysis of the results obtained and the symptoms observed in the original frontal and occipital experiments was performed by physicians [13]. They focused on the concussion output as measured by physiological changes and their possible correlation to pathological observations, including studies of hemorrhages, contusions and circulatory disturbances. Based on the presence of pathology in the brain stem and spinal cord, Kanda et al. [13] suggested that the physiological and pathological changes that took place in these regions were responsible for the concussions. This hypothesis was supported by the previous work with simulated occipital impacts [12] (Figure 10 in the appendix). However, it was uncertain if these observations could be extrapolated to simulated frontal impacts, which this study confirms. Moreover, the brain tissue responses observed in the simulations suggest that a rearward rotation of the skull may cause high brain stem strains based on two mechanisms. First, the rearward rotation of the skull induces a direct contact between the lower brain stem and the foramen magnum. Second, the cerebellum is pushed down by the tentorium, thereby compressing the brain stem.

Concussion caused by brain stem damage can be predicted in numerical simulations with injury threshold for strains. This study suggests that the threshold for strain in the brain stem for frontal impact is 0.27 MPS for 50% probability of injury. Assuming that the tissue level tolerances are equivalent for NHP and humans, this criterion can be directly applied to human FE models. Our findings are consistent with comparable simulation work in the literature in which sports accidents resulting in concussion were reconstructed with human FE models [27][11].

In the frontal impacts, all the global head injury criteria showed significantly higher values for the cases that scored concussion compared to those that did not. Analyzing their correlation to brain stem strain values, GAMBIT showed the highest correlation (0.82 for VMS and 0.81 for MPS). This seems reasonable since GAMBIT was designed to account for combined linear and rotational accelerations, and the simulated frontal impacts in our study are good examples of that, as shown in Figure 9. Thus, it may seem that GAMBIT is a good candidate to predict concussion caused by brain stem injury. However, GAMBIT cannot capture the underlying injury causation, as presented in the NHP experiments, since it is limited to global head accelerations, as illustrated by a simple example. If the resulting head accelerations from any one of the simulated cases is applied to prescribe the skull motion of a head-and-neck FE model where the vertebrae of cervical spine is rigidly attached to the skull, the resulting GAMBIT value will be the same while the brain stem strains will be much lower. For case 336, the peak MPS is reduced from 0.22 to 0.12 when the relative motion between the head and neck is removed, while the GAMBIT value is 2.2 for both cases. Complementary information on the duration of the applied rotational acceleration to the head may reduce this limitation of the GAMBIT criterion.

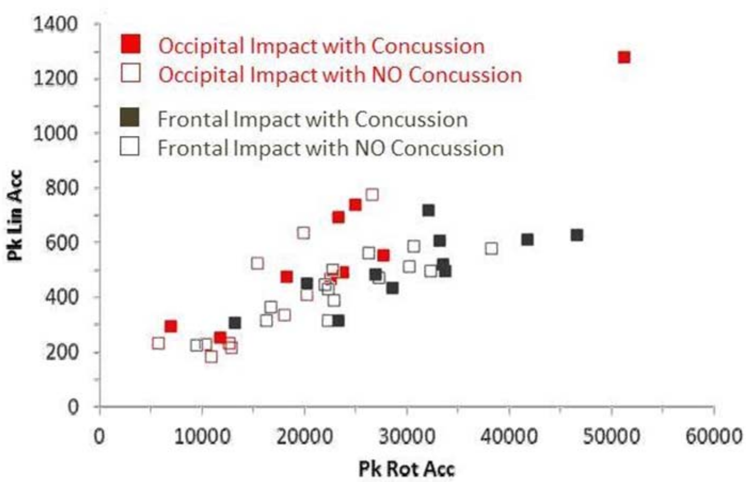


Fig. 9. Representation of peak linear acceleration (g) versus peak rotational acceleration (rad/s^2) as obtained from the frontal impact simulations presented in this study (green) and the occipital impact simulations in [12] (red). Occurrence of concussion (Filled: Concussion. Empty: No Concussion) for each experimental case is also represented.

The JARI experiments showed that the tolerance threshold for concussion, in terms of head linear accelerations, was higher for frontal impacts than occipital impacts [4]. Based on these observations, it may be suggested that the specimens used in the experiments were more sensitive to occipital impacts. In contrast, another NHP study by [1], which subjected 80 macaque specimens to rigid piston impacts, did not find that the thresholds varied significantly with the site of impact. Instead, the efficiency of the impact, i.e. if the piston impacted perpendicular to the surface or delivered a glancing blow, was identified as a critical issue. Simulation results from the current study, both global head injury criteria and brain stem strains, had tendencies of higher values for frontal impacts than for occipital impacts, consistent with [4]. However, the frontal impacts were more of a glancing type of impact than were the occipital impacts. In the occipital impacts, the energy was most likely transferred more efficiently to the scalp, the skull and the brain. To confirm this, the impactor impulse was calculated for a frontal impact (23N.s from the case in Figure 5) and an occipital impact with identical impact conditions (28N.s from the case in Figure 10), resulting in higher values for the occipital impact. Hence, in the occipital impact more energy is transferred to the head compared to the frontal impact under equivalent conditions. In other words, the differences in thresholds seen in the original experiments by [4] for frontal and occipital impacts and supported by the simulations here can be explained by differences in the impactor momentum transfer rather than higher injury thresholds for NHP in frontal impacts as compared to occipital impacts.

To conclude, support for the hypothesis of brain stem damage causing concussion and rejection of the hypothesis that brain injury thresholds depend on impact location indicate that frontal and occipital cases may be grouped together to study injury mechanism and evaluate brain injury criteria, thereby facilitating the development of protective strategies for concussion in saggital impacts.

VI. LIMITATIONS AND FUTURE WORK

The 24 frontal impacts simulated in this work were delivered to 8 specimens. The possible influence of accumulated brain damage in the specimens subjected to repeated impacts was not considered in this study. Nevertheless, if our results were affected by this, they would be affected towards conservative thresholds, since we are assuming that the injury thresholds of NHP subjected to multiple impacts would be the same as if they had not been subjected to impacts previously.

The same FE model limitations as described in [12] apply to this study. The brain tissue properties were validated against experimental data captured at low speeds, while the monkey FE model was used to simulate high speed impacts. In addition, the validation of the brain-skull relative motion is still limited. Moreover, the results and conclusions addressed in this study are restricted to the primary impact, since the technique used to simulate the impacts was limited to simulate the events during head contact and a short time (2 ms) after the contact. Additional simulation based sensitivity analysis will be conducted to clarify how these model related limitations may be affecting our results.

VII. CONCLUSIONS

In the simulated frontal impacts, the highest strains were seen in the brain stem tissue between the pons and the foramen magnum. This is consistent with physiological and pathological observations from the trauma experiments and to previous work with simulated occipital impacts.

Injury risk reference values for MPS in the brain stem are provided. These are intended to be used as reference values in the analysis of head impacts with human FE models.

Existing global head injury criteria were evaluated and compared. GAMBIT showed the highest correlation to brain stem strain. However, the criterion failed to capture the underlying mechanism that causes the brain stem strain and is thus rejected.

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IX. APPENDIX

Table V lists the experiment number (column 1, consistent with experiment numbering from [4]) impact conditions (columns 2 to 4) and the injury measurements including physiological changes and scored concussion (columns 5 to 8) from the former occipital impact experiments simulated in [12]. The table also includes the global head injury criteria (columns 9 to 12) newly calculated in this study and the brain tissue response (columns 13 to 16) from the simulations presented in [12].

TABLE V

OCCIPITAL IMPACT TEST CONDITIONS, INJURY OUTPUT AND SIMULATION RESULTS.

	From experiments							From simulations							
	Impact conditions			Injury measurements				Global head injury criteria				Brain tissue parameters			
												Cerebrum		Brain stem	
Exp No	Impact velocity (m/s)	Rubber block (1)	Impact Stroke (mm)	Apnea (s)	Blink (s)	Brady cardia (2)	Concussion (Grade) (3)	HIC	GAMBIT	BRIC	RIC	VMS (Pa)	MPS	VMS (Pa)	MPS
311	10.9	B	20	0	16	No	No (0)	4648	2.0	1.3	2.4E+8	4172	0.32	4193	0.20
312	14.4	B	20	0	18	No	No (0)	6995	2.5	1.7	3.2E+8	4573	0.35	4803	0.22
313	16.3	B	20	0	30	No	Yes (I)	8496	2.8	1.9	4.0E+8	4809	0.37	5111	0.24
314	18.0	B	20	0	34	No	Yes (I)	10130	3.0	2.1	4.4E+8	5024	0.38	5590	0.25
316	19.8	B	20	0	11	No	No (0)	11968	3.1	2.2	4.9E+8	5152	0.40	5818	0.27
321	21.7	A	30	36	12	Yes	Yes (II)	60233	5.6	4.3	2.3E+9	8549	0.65	7365	0.34
322	15.5	D	30	32	13	No	Yes (I)	8205	2.0	1.5	3.9E+8	4297	0.33	3849	0.18
344	10.8	E	40	0	5	No	No (0)	4965	1.8	1.5	3.9E+8	3543	0.28	3255	0.15
345	17.0	E	40	0	30	No	Yes (I)	9539	2.3	1.9	3.8E+8	4265	0.33	4200	0.19
356	6.8	E	60	0	19	No	No (0)	2158	1.1	0.9	1.4E+8	3298	0.26	3583	0.16
357	7.5	E	90	10	33	No	Yes (I)	2787	1.2	1.0	1.8E+8	3684	0.29	3648	0.15
358	11.5	E	90	0	18	No	No (0)	8330	2.1	1.7	5.5E+8	4774	0.37	4287	0.19
359	13.7	E	60	0	30	Yes	Yes (II)	12358	2.4	2.0	8.3E+8	5186	0.39	4708	0.21
360	16.4	E	60	20	14	Yes	Yes (II)	16112	2.8	2.3	9.6E+8	5165	0.39	4937	0.22
362	10.8	E	10	0	10	No	No (0)	559	0.8	0.5	1.9E+7	2368	0.17	2705	0.12
363	13.4	E	10	19	10	Yes	Yes (I)	958	1.0	0.6	3.8E+7	2759	0.22	3125	0.14
370	8.0	E	30	0	11	No	No (0)	2031	1.3	1.1	1.8E+8	3273	0.26	3073	0.16
371	7.3	E	20	0	6	No	No (0)	945	1.0	0.9	6.8E+7	2221	0.19	2535	0.12
372	8.5	E	25	0	10	No	No (0)	1655	1.2	1.1	1.2E+8	3090	0.24	2878	0.13

(1) Impactor rubber block type according to [4].
(2) Bradycardia 'Yes' when two levels of blood pressure disturbance followed impact.
(3) Occurrence of concussion used in [12]. In brackets, the concussion grade scored in the experiments

Figure 10 illustrates the head kinematics and brain strains for one simulated occipital impact with the same impact conditions (Impact speed 20.1m/s, intermediate stiffness (C), and maximum stroke 45mm) as the frontal impact case analyzed in Figure 5.

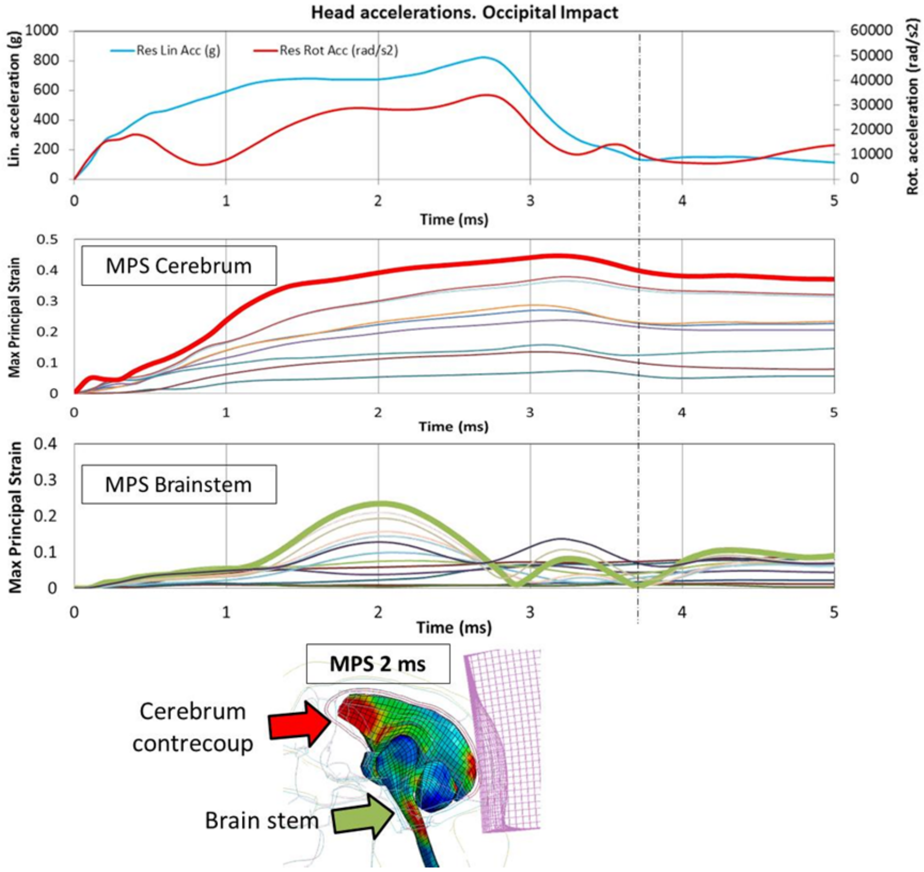

Fig. 10. Analysis of head kinematics in relation to brain tissue strains in a virtual occipital impact. Head linear and rotational acceleration during impact (top figure), maximum principal strain curves of several elements in the cerebrum and the brain stem (middle figures) including the one that showed the highest values (in red and green for the cerebrum and the brain stem, respectively) and fringe plots of the maximum principal strains in the brain at 2 ms.