Injury tolerances for oblique impact helmet testing

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Abstract: The most frequently sustained severe injuries in motorcycle crashes are injuries to the head, and many of these are caused by rotational force. Rotational force is most commonly the result of oblique impacts to the head. Good testing methods for evaluating the effects of such impacts are currently lacking. There is also a need for improving our understanding of the effects of oblique impacts on the human head. Helmet standards currently in use today do not measure rotational effects in test dummy heads. However rotational force to the head results in large shear strains arising in the brain, which has been proposed as a cause of traumatic brain injuries like diffuse axonal injuries (DAI).

This paper investigates a number of well-defined impacts, simulated using a detailed finite element (FE) model of the human head, an FE model of the Hybrid III dummy head and an FE model of a helmet. The same simulations were performed on both the FE human head model and the FE Hybrid III head model, both fitted with helmets. Simulations on both these heads were performed to describe the relationship between load levels in the FE Hybrid III head model and strains in the brain tissue in the FE human head model. In this study, the change in rotational velocity and the head injury criterion (HIC) value were chosen as appropriate measurements. It was concluded that both rotational and translational effects are important when predicting the strain levels in the human brain.

Keywords: Injury tolerances, Oblique impacts, Helmet, Head injuries and FE (Finite Element)

INTRODUCTION

Motorcycle riders are very exposed in traffic accidents since their level of protection is limited. One way to provide better safety for motorcycle riders is to improve the helmets. The most frequently sustained severe injuries in motorcycle crashes are namely injuries to the head [1]. Many of these injuries are caused by rotational forces [2] that are most commonly generated as a result of oblique impacts found in motorcycle crashes [3]. The most frequent impact is an impact close to the visor attachment points with an average impact thereof speed of 44 km/h and an average angle of 28° to the impacted object [3].

Most safety helmets are designed to meet the requirements established by standardised tests (e.g. ECE22.05 [4], FMVSS218 [5], and BS6658 [6]). In these tests, the helmet is typically dropped vertically onto a flat or curved rigid surface that is set up for a tangential impact to the helmet surface. During the drop tests, the translational acceleration of a head form is measured. The British Standard BS6658 includes an oblique impact, but there is no measurement of the rotational effects, which makes it difficult to correlate these effects to injuries in the human brain. The oblique impact tests in the BS 6658 ensure:

- that projecting visor mounts and other projections shear off easily when there is an impact with a series of parallel bars; and
- that the tangential force on the helmet shell, when it impacts with a rough flat surface, is no larger than that of typical shell materials used in 1985 (the year of introduction of the test).

The ECE22.05 standard also includes a test that can be seen as an oblique impact test. However, this is a test for projection and surface friction, and does not include rotational measurements in the dummy head. Today's standards have resulted in helmets with a good degree of protection for radial impacts while their protection for oblique impacts remains unknown.

In an oblique impact to a helmeted head, the rotational energy can be absorbed by (a) frictional energy between

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the helmet shell and the impacting surface, (b) internal energy in the liner and comfort foam due to shear deformation, and (c) frictional energy between the liner and the head.

The standard test procedures are based on vary vague reasons. More realistic would be to test helmets for impacts similar to the most commonly observed impacts in real life motorcycle accidents. Aare and Halldin (2003) [7] proposed a new method to test helmets for oblique impacts, but they did not propose any injury tolerances for such a test. The injury tolerances and criteria used when testing oblique impacts to helmets should predict injuries frequently seen in traffic accidents. The most frequently sustained brain injuries in motor vehicle accidents that result in either fatality or the need for long-term rehabilitation are subdural hematomas (SDH) and diffuse axonal injuries (DAI) [2]. The main causes of SDH are ruptured arteries or bridging veins. DAI is caused by the tearing of neuronal axons in the brain tissue. It has been proposed that there are correlations between these injuries and rotational effects to the head [2]; [8]. In these studies injury thresholds for purely angular motions was proposed. If an angular motion is combined with a translational motion, these thresholds probably have to be decreased, as shown in the study by DiMasi et al. (1995) [9] and Ueno and Melvin (1995) [10].

To establish tolerances or criteria for an oblique impact to the helmeted head it is possible to use a detailed finite element (FE) model of the human head [11]. This method should be seen as a complement to biological experiments on human cadavers. It has been suggested that there is a correlation between the strain in the brain and DAI [12]; [13]. If the FE model of the human head is well correlated with relevant experiments, then the strain computed in the model can be compared with accepted tissue-based injury thresholds. Bain and Meaney (2000) [12] proposed a threshold of 20% strain in the brain tissue for the onset of the malfunction of the nerves in the brain, which could be seen as a first stage of DAI.

Currently there is a lack of good testing methods for evaluating the effects of oblique impacts. There is also a need for improved understanding of the effects on the human head subjected to oblique impacts. In current helmet standards, no rotational effects are measured in the head form, partly because there are no accepted global injury thresholds for a combination of rotations and translations. The objective of this study was therefore to study if, and how, rotational and translational parameters influence the strain levels in the brain for three well-defined and commonly observed [3] oblique impacts to helmets.

The aim of this study was to propose new injury tolerances for a specific set of oblique helmet impacts. Further it was hypothesized that it is possible to predict the strain in brain tissue using the peak change of rotational velocity and the HIC value.

METHODS

This is a numerical study using the non-linear and dynamic finite element (FE) code LS-DYNA [14]. In this study, FE models of (1) the human head [15], (2) a Hybrid III dummy head [16], and (3) an experimental helmet were used. Simulations where performed on both the human head and on the Hybrid III dummy head, wearing the helmet.

Injury tolerances

The maximal principal strain in the brain tissue was chosen as a predictor of injuries, as it has been shown to correlate with DAI [12]. A strain of 20% was shown to be critical to the brain tissue. As the strains in the brain tissue are proportional to the HIC value for pure translations, and also proportional to the change in rotational velocity for pure rotations of short impact durations [11], it is suggested that the output data is fitted to the following formula:

$$\varepsilon = k_1\Delta\omega + k_2 HIC \qquad [1]$$

where ε is the maximum strain in the brain tissue, $\Delta\omega$ is the peak resultant change in rotational velocity, HIC is the head injury criteria [17] and k_1 and k_2 are constants. The strain (ε) is taken from the FE human head model, whereas $\Delta\omega$ and HIC are taken from FE Hybrid III head model.

FE model of the human head

The FE model of the human head (Figure 1) developed at the Royal Institute of Technology (KTH) Stockholm, Sweden, is based on data from the Visual Human Database

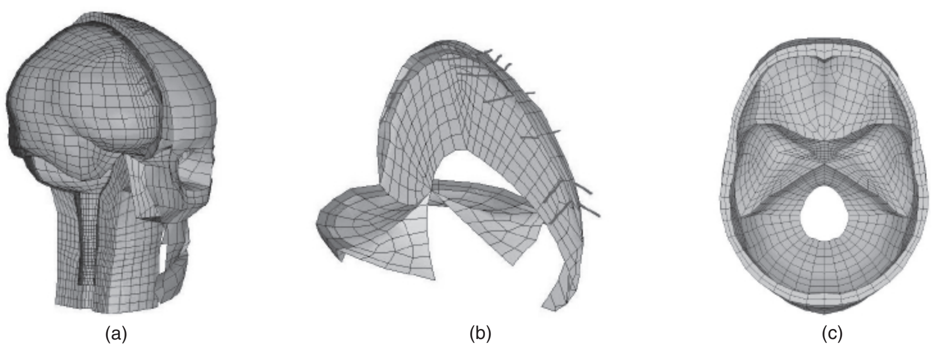

Figure 1 Finite element mesh of (a) the human head, (b) falx and tentorium including transverse and superior sagittal sinuses with bridging veins, and (c) cranium.

[18]. The FE model is anatomically detailed and includes the scalp, the skull, the brain, the meninges, the cerebrospinal fluid (CSF) and eleven pairs of parasagittal bridging veins [15], [19], [20] and [21].

FE model of the Hybrid III dummy head

The FE model of the Hybrid III dummy head is very similar to the real Hybrid III dummy, with respect to geometry, mass, inertia, and material properties [16]. The model used in this study does not include the neck. This is a simplification used in all standard helmet tests. The solid part of the head (the aluminium) was modelled as rigid, whereas the rubber skin was modelled using a viscoelastic material model [16]. The Hybrid III head model used in this study was meshed using 2311 elements (two elements through the thickness in the solid part making a total of 1288 elements and two elements through the thickness in the rubber skin making a total of 1023 elements).

FE model of the helmet

The FE model of the helmet includes a linearly-elastic material model of the ABS thermoplastic shell, validated against quasi-static tensile tests (thickness: 4.7 mm, Young's modulus: 1.64 GPa, density: 1161 kg/m^3 and Poisson's ratio: 0.45). The energy absorbing liner material consists of expanded polystyrene (EPS) with a density of 40 kg/m^3 modelled using a crushable foam material [14]. This foam material model requires a stress-strain curve. Therefore hydrostatic compressive tests were performed on expanded polystyrene (Figure 2). The data from this test were implemented in the material model. Poisson's ratio was estimated to zero and Young's modulus implemented in the material model was estimated to 8 MPa. Young's modulus implemented in this material model represents the stiffness during unloading. The stiffness during loading is taken from the implemented load-curve. The tensile stress cut-off was estimated at 1 MPa and the damping coefficient estimated at 0.05 since the material was assumed not to be strain-rate dependent [22].

The helmet model was not fitted with a chinstrap, as it was concluded when studying high-speed movies from experiments performed by Aare M. and Halldin P [7] that the inclusion of a chin strap made no significant difference to well-fitting helmets during short duration impacts.

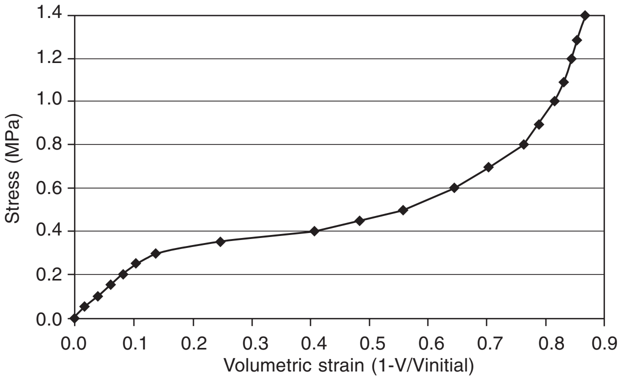

Figure 2 Stress-strain curve from the hydrostatic test.

The contact definition between (a) the FE model of the human head and the helmet and (b) the FE model of the Hybrid III dummy head and the helmet was the "surface-to-surface interference" [14]. This means that when the head is larger than the space inside the helmet, an initial pressure or contact force arises. When the head is smaller than the space inside the helmet, there is no initial pressure. However, the inside surface of the helmet fitted with the Hybrid III head model was shaped to fit the geometry of the head perfectly. The inside surface of the helmet fitted with the human head model was shaped to fit the geometry of that head perfectly. This means that there was no initial gap between the heads and the helmets anywhere throughout the contact surfaces. This also means that the geometry of the inside surface if the helmets were slightly different. However, the thickness of the liner in the two different helmets was similar.

FE model of the helmet and the Hybrid III head combined

The complete model of the helmet and the Hybrid III dummy was validated against both radial and oblique impact tests on helmets (Figure 3). Some of these experimental data was presented by Aare and Halldin (2003) [7].

Numerical simulations

Three different impacts were tested, where impact 1 is to the top of the head inducing sagital plane rotation, Impact 2 is lateral inducing axial rotation, and Impact 3 is lateral inducing coronal plane rotation (Figure 3). For Impacts 1 and 3, three different impact velocities were used (5, 7 and 9 m/s), and for Impact 2 an additional impact velocity of 3 m/s was also used in the testing. The reason for simulating impact 2 at 3 m/s was that all the other impact velocities caused strains in the brain tissue larger than 20%. The impact angles were induced by altering the speed of the head and the speed of the impactor (Table 1). For all three impacts and impact velocities, four different impact angles between the head and the impactor were tested (30°, 45°, 60° and 90°). These specific scenarios were chosen to cover the range of the most commonly observed impacts in real life motorcycle accidents [3].

Data from the helmeted Hybrid III head was correlated to load levels in the human brain, by performing simulations on both the FE human head and the FE Hybrid III head. Comparisons were made between the strains in the brain tissue of the FE human head and the change of rotational velocity and the HIC-value in the FE Hybrid III head for identical impacts. These comparisons were done to find the relationship between the global kinematics of a Hybrid III head and the strains in the human brain.

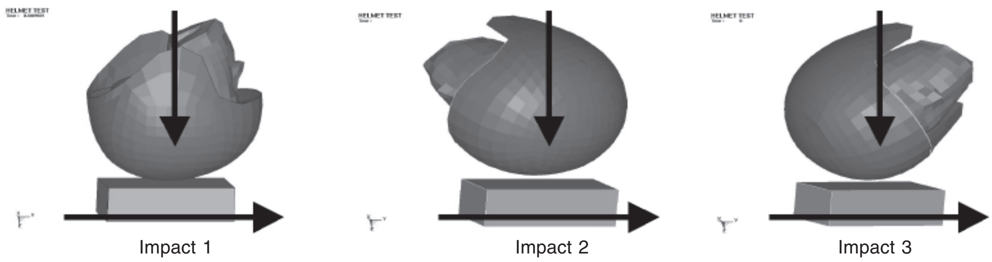

Figure 3 The three different impacts in the numerical simulations.

Table 1 **Vertical and horizontal velocities of the different impact speeds and angles used in the numerical simulations.**

	Impact velocity: **3 m/s**		Impact velocity: **5 m/s**		Impact velocity: **7 m/s**		Impact velocity: **9 m/s**	
Angle of impact	Vertical impact speed (m/s)	Horizontal speed of the plate (m/s)	Vertical impact speed (m/s)	Horizontal speed of the plate (m/s)	Vertical impact speed (m/s)	Horizontal speed of the plate (m/s)	Vertical impact speed (m/s)	Horizontal speed of the plate (m/s)
30°	1.50	2.60	2.50	4.33	3.50	6.06	4.50	7.79
45°	2.12	2.12	3.54	3.54	4.95	4.95	6.36	6.36
60°	2.60	1.50	4.33	2.50	6.06	3.50	7.79	4.50
90°	3.00	0.00	5.00	0.00	7.00	0.00	9.00	0.00

RESULTS

Figure 4 shows a comparison between experiments and the numerical simulations from impact 1, velocity 7 m/s and impact angle 45°.

The results from the numerical simulations are presented in Table 2.

Comparing the results from the various different impacts, it is quite clear that the rotational effects have a major influence on the strain levels in the human brain (Figures 5, 6 and 7). Each point in these figures represents one simulated impact. These points where plotted to give the reader a better picture of which impact scenarios that are critical. The maximum strain in the brain was usually found in the white brain tissue, though in different elements. The constants k_1 and k_2 in Equation 1 were computed for Impacts 1, 2 and 3 respectively, using least square regression analyses of Equation 1 (Table 3). The regression coefficient is a figure that describes how well the data correlates to the equation. A regression coefficient of one represents a perfect match and a regression coefficient of zero represents no correlation at all.

Different isostrain curves can be plotted using the computed constants k_1 and k_2 in Equation 1 (Figures 5, 6 and 7).

DISCUSSION

This study has presented a method for computing test-specific injury tolerances using an FE model of the human head. The two regression parameters, HIC and change in rotational velocity, vary in their influence on the head response depending on the impact location and orientation of the helmeted head (Figures 5, 6 and 7). In Impact 1, the change in rotational velocity exhibits a fairly sharp cut-off point for determining safe and unsafe conditions, whereas in Impact 2 and 3, the HIC value (or at least a translational component) also needs to be considered when predicting injury. It is therefore difficult to use the one injury tolerance indicator for all types of impacts. It is suggested that one specific set of constants be used for each impact direction and speed.

Regression analysis was used in this study to fit the data to equation 1. The regression coefficient is a figure that describes how well the data correlates to the equation. A regression coefficient of one represents a perfect match and a regression coefficient of zero represents no correlation at all. The regression analysis showed good correlations for Impacts 1 and 2 (Table 2). For this impact, a formula of higher order would probably fit the data better, especially for very low changes in rotational velocity. It is likely that the curves plotted in figures 5, 6 and 7 are more accurate in the central parts of the diagrams close to the load levels that are used in the simulations.

The simulated impacts in this study were not only chosen to cover the most commonly observed impacts in motorcycle accidents, but also to cover rotations around all three Cartesian axes. Impact 2 was according to statistics, the most common impact in real-life accidents.

The head is vulnerable to accelerations in the lateral direction [23], [24] and therefore the HIC values are more critical for Impacts 2 and 3 than for Impact 1.

Even when the head was dropped vertically (90°), rotations were induced in all the impacts and in particular in Impacts 2 and 3. These rotations occur because the impact point is not situated directly under the centre of gravity.

For rotational impulses of short duration, the change

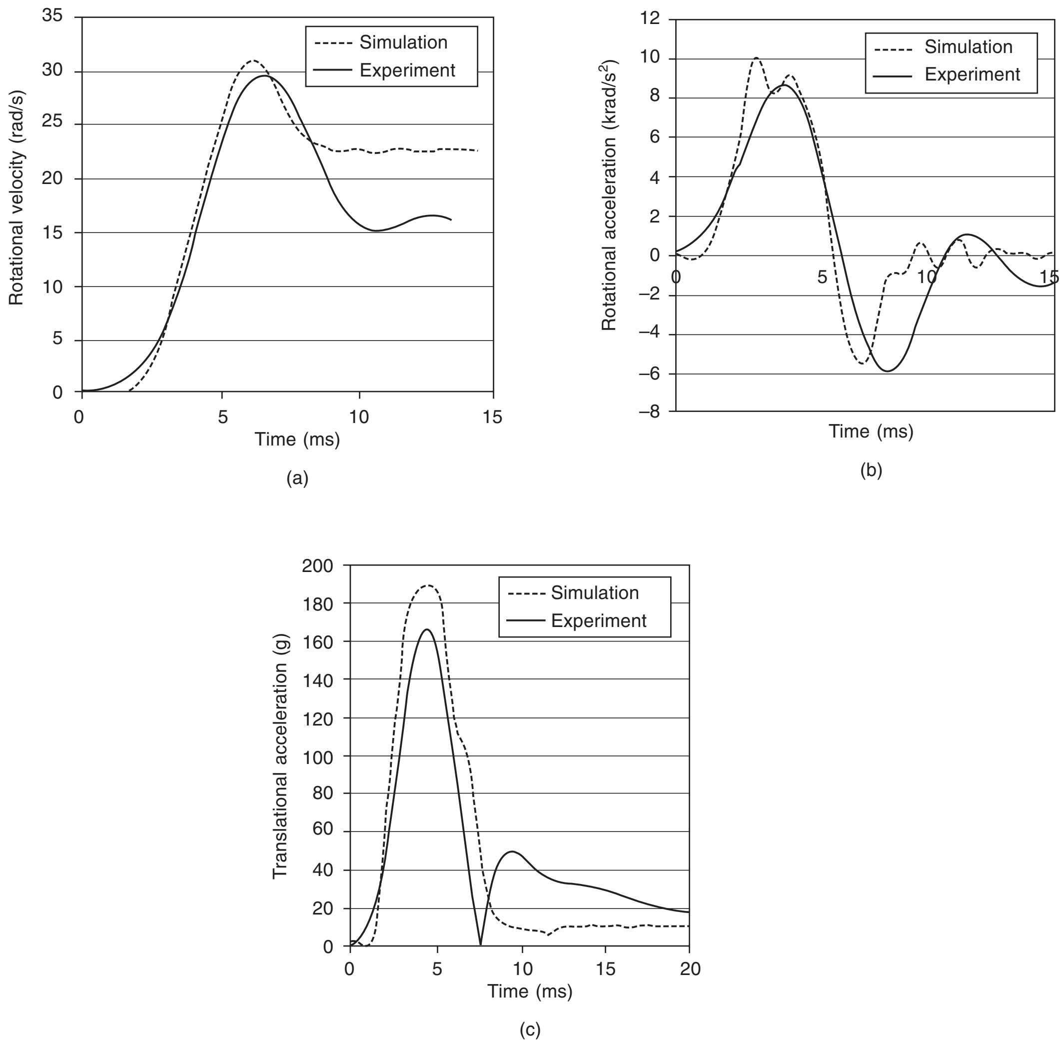

Figure 4 Comparison of simulations and experiments from impact 1 (Figure 3), velocity 7 m/s and impact angle 45°, where (a) is the change of rotational velocity, (b) is rotational acceleration, and (c) is translational acceleration.

in angular velocity has been shown to correspond better than all other parameters with the intracranial strains found in the FE model [11]. This is in agreement with Holbourn's hypothesis [13]. For translational impulses on the other hand, the HIC [17] and the HIP [25] have shown the best correlation with the strain levels found in the model [11].

In Table 2 all data needed to plot Figures 5, 6 and 7 are displayed. Additionally, the rotational accelerations are displayed in Table 2, since these parameters are used in some other injury criteria [8] and might therefore be interesting to some readers. The translational acceleration is also displayed in Table 2, since this parameter is correlated to the HIC-value and might therefore also be of interest.

An FE model of the human head was used to compute the maximal principal strain in the brain, and thereby analyse the risk for DAI. There are also other tissue-level measurements that may be used as injury predictors, such as strain rate, the product of strain and strain rate, von-Mises stress, and strain energy. In this study, the strain has been chosen to analyse the risk for DAI, as this measurement has been experimentally verified previously [12], [26]. An FE model needs to be well correlated to relevant experimental studies on the human brain. Another important issue in modelling of the human head is the selection of appropriate material properties for various intracranial structures. The choice of shear properties for the brain tissue is difficult as the range of published values varies several orders of magnitude [21]; [27]. However, FE model used in this study has been carefully validated and shown to have a good correlation with experiments found in the literature including rotational injuries correlated to strain in the brain as well as local brain motion experiments [28], [29].

Another limitation with the test method presented here (and indeed with most helmet testing methods) is that only the head is used. A more realistic simulation would probably involve using the whole body, or at least the head, neck and torso. Involving for example the neck would change the dynamics of the impact, as the boundary conditions for the head would change. Ruan et. al. (1991) [30] showed that a single hinge coupling between the head and the neck had a limited effect on the intra-cranial

Table 2 **Results from the numerical simulations.**

			FE model of the Hybrid III head				FE model of the human head
Impact area*	Impact velocity	Impact angle	Δω (rad/s)	HIC-value	Peak resultant translational acc (g)	Peak resultant rotational acc (krad/s^2)	Maximum strain in brain tissue
1	5	30	26.6	320	102	9.1	0.16
1	5	45	21.7	651	132	8.5	0.13
1	5	60	12.2	974	165	4.6	0.08
1	5	90	6.8	1183	191	2.6	0.10
1	7	30	35.6	652	145	11.8	0.23
1	7	45	26.5	1363	188	7.8	0.19
1	7	60	14.6	1939	217	4.9	0.11
1	7	90	9.5	2279	243	3.6	0.12
1	9	30	45.0	1083	177	14.2	0.30
1	9	45	34.4	2206	220	11.6	0.24
1	9	60	21.2	3178	264	9.2	0.14
1	9	90	12.3	3791	294	4.8	0.12
2	3	30	20.8	50	52	6.3	0.13
2	3	45	19.5	95	65	6.5	0.13
2	3	60	16.9	139	75	5.9	0.14
2	3	90	12.9	183	80	4.7	0.12
2	5	30	32.9	134	73	9.2	0.22
2	5	45	31.3	262	97	10.0	0.23
2	5	60	27.4	394	115	9.4	0.23
2	5	90	21.5	536	126	8.8	0.19
2	7	30	44.4	255	92	11.4	0.32
2	7	45	42.7	515	126	12.9	0.32
2	7	60	36.9	811	151	12.3	0.31
2	7	90	29.0	1252	165	11.6	0.25
2	9	30	54.1	384	109	13.4	0.40
2	9	45	52.8	883	152	16.1	0.43
2	9	60	46.3	1658	196	15.0	0.43
2	9	90	34.3	3152	280	13.4	0.34
3	5	30	25.2	318	100	7.4	0.12
3	5	45	16.6	614	133	8.4	0.14
3	5	60	7.3	827	157	2.9	0.15
3	5	90	15.3	922	171	5.6	0.19
3	7	30	33.7	664	134	9.9	0.18
3	7	45	22.3	1244	173	8.7	0.18
3	7	60	12.2	1670	195	6.8	0.20
3	7	90	20.0	1840	220	8.6	0.25
3	9	30	43.6	1094	162	12.0	0.24
3	9	45	30.4	2119	210	8.1	0.23
3	9	60	16.9	3010	239	9.8	0.27
3	9	90	24.8	3553	285	10.0	0.34

* Defined in figure 3

pressure during impact. Beusenberg et. al. (2001) [31] simulated the influence of different neck models using data from head impacts recorded from the National Football League in the US. In that study, it was shown that for oblique impacts to the front of the head, the coupling between the head and the neck is only important for rotations in the sagital plane. These rotations are however strongly dependent on the boundary conditions of the neck. Hering and Derler (2000) [32] performed both radial and oblique helmet impact tests on both a detached Hybrid III dummy head and a complete Hybrid III dummy. It could be concluded from their study, that the influence of the neck and body on the rotational effects in the head was different for different impact locations. They also

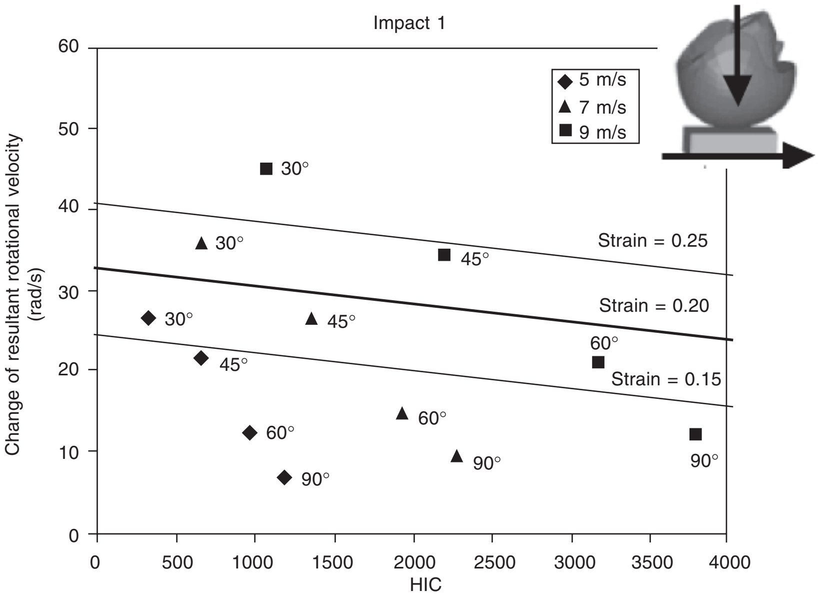

Figure 5 Results from impact 1, where the axes represents computed levels in the FE Hybrid III head model, and the straight lines are isostrain curves representing different strain levels in the brain tissue in the FE human head model.

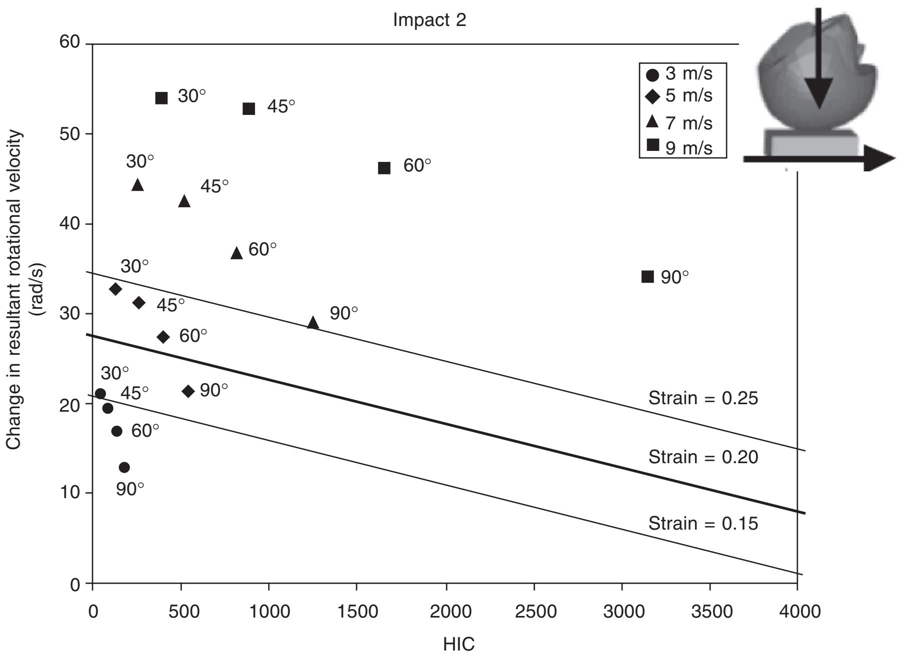

Figure 6 Results from impact 2, where the axes represents computed levels in the FE Hybrid III head model, and the straight lines are isostrain curves representing different strain levels in the brain tissue in the FE human head model.

concluded that because the Hybrid III neck is much stiffer than the human neck, this presented a problem. It is proposed here that the influence of the neck ought to be investigated in future studies.

CONCLUSIONS

When comparing the FE Hybrid III head model kinematics with strains in the FE human brain tissue during oblique impacts, it can be concluded that rotational kinematics are as important as translational kinematics and should therefore be included in future head injury criteria. In Impact 1, changes in the rotational velocity provide a critical parameter. In Impact 2 and 3, changes in rotational velocity as well as the HIC value are important indicators (reiterating here, Impact 1 is to the top of the head inducing sagital plane rotation, Impact 2 is lateral inducing axial rotation, and Impact 3 is lateral inducing coronal plane rotation).

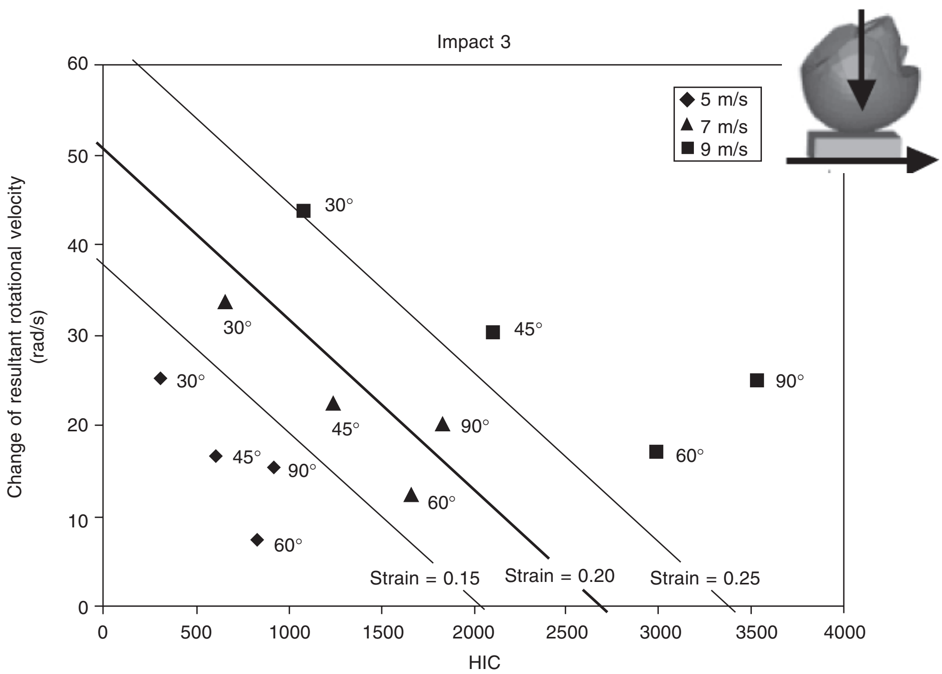

Figure 7 Results from impact 3, where the axes represents computed levels in the FE Hybrid III head model, and the straight lines are isostrain curves representing different strain levels in the brain tissue in the FE human head model.

Table 3 **Constants k_1 and k_2 and the regression coefficient r^2 from the regressions analysis.**

	Impact 1	Impact 2	Impact 3
k_1	$6.14*10^{-3}$	$7.26*10^{-3}$	$3.92*10^{-3}$
k_2	$1.32*10^{-5}$	$3.50*10^{-5}$	$7.41*10^{-5}$
r^2	0.93	0.97	0.68

This study shows that it is possible to make a good prediction of the strain in the brain tissue using the peak change of rotational velocity and the HIC value. Test-specific injury thresholds should therefore include both rotational and translational parameters, such as the change in rotational velocity and HIC. The results presented here can be helpful in preventing head injuries through their consideration in the setting of future standards for oblique impact helmet tests.

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