A PARAMETRIC STUDY OF ENERGY ABSORBING FOAMS FOR HEAD INJURY PREVENTION

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ABSTRACT

This paper describes a parametric study of foam material properties for interior car surfaces using finite element calculations. Two different head models were used for the impact simulations, a Hybrid III dummy head and a biomechanical head model. The objective was to study the head injury criterion (dummy) (HIC(d)), the angular velocity, the resultant acceleration and, for the human head models, the strain in the brain tissue and the stress in the skull for a variation in foam material properties such as stiffness, plateau stress and energy absorption. The analysis gave at hand that the best choice of material properties with respect to impact using the Hybrid III head model reached different results compared to an impact with the biomechanical head model. For a purely perpendicular impact, the HIC(d) for the head model managed to predict the strain level in the brain quite well. Even though the HIC reached acceptable levels for both a perpendicular and oblique impact towards a 31 kg/m^3 EPP padding, the maximum strain in the human head model for an oblique impact was almost twice suggested allowable levels. The difference in the strain in the brain between an oblique and perpendicular impact when impacted with same initial velocity towards the same padding was not predicted by the HIC(d).

INTRODUCTION

Head injuries due to traffic accidents, at work and during leisure, are major diseases in Sweden and worldwide. Globally, the daily incidence rate of transportation injuries is estimated to 30 000 victims and 3 000 deaths [1]. In Sweden, the annual number of cases is more than 20 000 head injuries and the annual rate of head injuries in Sweden over the last 14 years is relatively constant [2]. The main cause of death for people younger than 45 years of age in Sweden is accidents and poisoning. When looking deeper into this cause of death for the younger part of the male population in Sweden, it can be seen that head injuries causes almost 80 percent of the traffic injury deaths [3]. The development of safety systems in cars has exploded over the last 20 years, resulting in more and more sophisticated methodologies. There are indications that this trend is slowing down. One possible factor is that the crash dummies are not completely human-like and another factor is the roughness of the tolerances and injury criteria that are used to couple output from the dummies with real-life injuries. The interior surfaces of a car compartment are designed to protect the occupants from injury at car accidents through use of energy absorbing materials and clever structural solutions. This is normally done to comply with the extended FMVSS 201 regulation [4]. The primary verification tool in the design process is the Head Injury Criterion (dummy) (HIC(d)) applied in a free motion head-form experimental set-up, where a rigid dummy head is launched towards specific locations. Linear accelerations in three perpendicular directions are measured in the head form during the impact and the performance is evaluated according to the HIC. The test procedure is established internationally and thus used by automotive manufacturers all over the world. HIC was introduced in its present form in crash testing by the National Highway Traffic Society Administration [5] and it has been used for several years in crash injury research and prevention as a measure of the likelihood of serious brain injury. HIC only treats the resultant translational acceleration and the duration of the impulse and no consideration is given to the direction of the impulse or rotational acceleration components [3, 6, 7]. Moreover, studies by Ueno and Melvin [8] and DiMasi *et al.* [9] found that the use of either translation or rotation alone may underestimate the severity of an injury. Zhang *et al.* [10] also concluded that both linear and angular accelerations are significant causes of mild traumatic brain injuries. Recently, it was found that HIC manage to predict the strain level in the brain of a finite element (FE) model for purely translational impulses of short duration, while the peak change in angular velocity showed the best correlation with the strain levels in an FE head model for purely rotational impulses [3]. The HIC(d) together with FE simulations and/or experiments according to the FMVSS 201 regulation has been used in several studies in an effort to improve the interior safety of

vehicles [11, 12, 13, 14]. However, the human head behaves in a more complex way and since the validity of the HIC criterion is intensively debated there is reason to believe that the safety development could be made more efficient through use of more delicate tools in the process, such as biomechanically representative FE models of the human head together with local tissue thresholds. To ensure that a continued high pace is kept when it comes to progress in car safety and primary prevention, it is necessary to find new preventive strategies and methods to complement the safety work practiced today. It is hypothesized in this study that the best choice of parameters for energy absorbing foams of an automotive panel would come out differently if it was made with respect to one or the other criterion. To test this hypothesis, different head models were compared in FE simulations according to the FMVSS 201 regulation using a simplified interior padding. This investigation was performed to illustrate that although the response of a structure may be optimal for a certain impact case when evaluated with a specific set of criteria it might not be favorable for another case, evaluated with respect to another set of criteria.

METHODOLOGY

In order to investigate the potential to improve the safety design, an FE model of the human head has been used. Two different FE head models were used; a model of the featureless Hybrid III dummy head and a biomechanically representative human head model (in the following referred to as human head model). Parametric studies of material properties of energy absorbing foams for idealized impact paddings were performed. Numerical simulations using the dynamic finite element method (FEM) program LS-DYNA [15] was performed.

Human head FE model

The head model used in this study was developed at the Royal Institute of Technology in Stockholm [16]. The head model includes the scalp, the skull, the brain, the meninges, the cerebrospinal fluid (CSF) and eleven pairs of the largest parasagittal bridging veins (Figure 1).

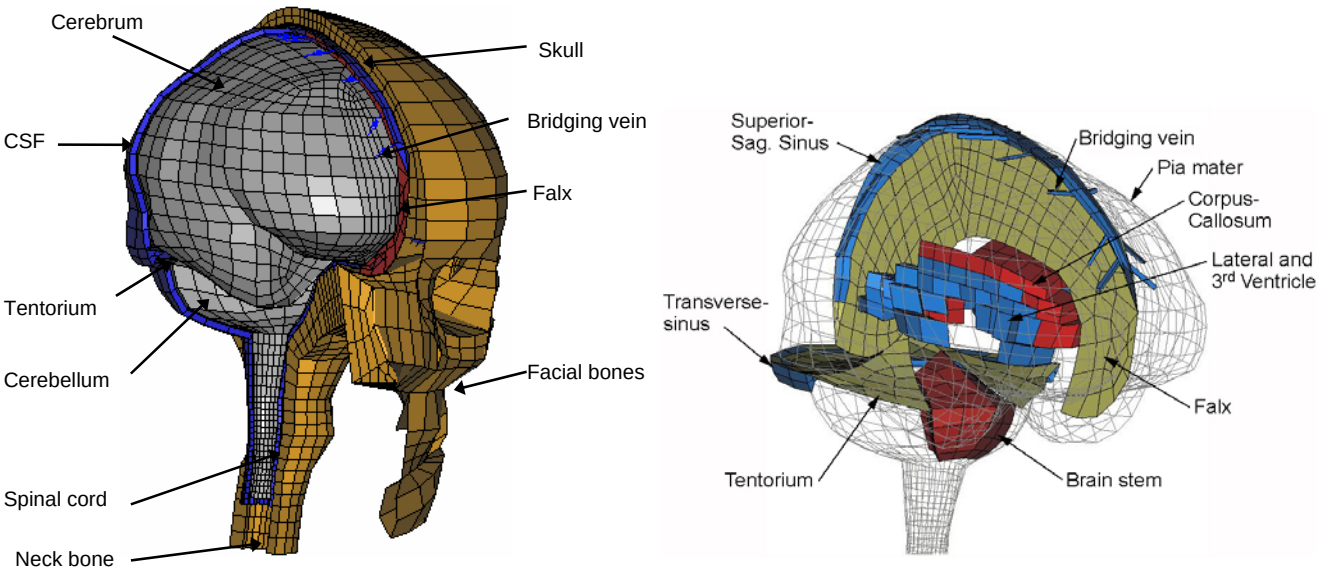


Figure 1. Finite element model of the human head.

In order to better simulate the stress and strain distribution, separate representations of gray and white matter, and inclusion of the ventricles were implemented. The total mass of the head was 4.52 kg and the principal mass moments of inertia were close to the corresponding ones for the hybrid III head. The head model has been validated against several relative motion experiments [17], intra-cerebral acceleration experiments [3], skull fracture experiments [18], and intra-cranial pressure experiments [19]. The post-mortem human subject (PMHS) experimental data used cover four impact directions (frontal, occipital, lateral and axial), short and long durational impacts (2-150 ms), high and low severity (sub-concussive to lethal), and both penetrating and non-penetrating injuries. To cope

with the large elastic deformations, a third order Ogden hyperelastic constitutive model and corresponding parameters was fitted to include the non-linear elasticity described by Miller and Chinzei [20] as well as the high frequency relaxation modulii determined by Nicolle *et al.* [21]. The stress in the cranial bone, maximum principal strain in the brain tissue, change in rotational velocity of the skull, the HIC(d) and translational acceleration of the skull for the different foams were determined. To account for the possible loss of load bearing capacity at high contact loading, the stresses in the skull were limited to 90 MPa for the compact bone [22, 23, 24] and 30 MPa for the spongeous bone [22, 25] through the use of simple elastic ideally plastic constitutive models. A summary of the properties for the tissues of the human head used in this study is presented in Table 1.

Table 1.
Material properties for the head model used in the numerical study.

Tissue	*Young's modulus* [MPa]	*Density* [kg/dm^3]	*Poisson's ratio*	*Yield stress* [MPa]
Outer compact bone	15 000	2.00	0.22	90
Inner compact bone	15 000	2.00	0.22	90
Porous bone	1000	1.30	0.24	30
Neck bone	1000	1.30	0.24	
Brain	Hyper-Viscoelastic	1.04	~0.5	
Cerebrospinal Fluid	K = 2.1 GPa	1.00	0.5	
Sinuses	K = 2.1 GPa	1.00	0.5	
Dura mater	31.5	1.13	0.45	
Falx/Tentorium	31.5	1.13	0.45	
Scalp	Viscoelastic	1.13	0.42	
Bridging veins	EA = 1.9 N			

K = Bulk modulus, and EA = Force/unit strain.

FE Hybrid III dummy head

The FE Hybrid III 50th percentile dummy head developed by Fredriksson [26], Figure 2, comprises a rigid skull covered in rubber flesh. The rubber was modeled using material properties according to the calibration tests by Fredriksson [26]. The total weight of the head was 4.52 kg. For stability reasons the head was made featureless by suppression of the nose.

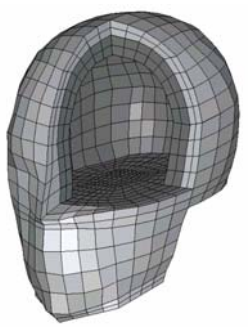


Figure 2. Finite element model of the featureless Hybrid III dummy head.

FE calculations

According to the FMVSS 201 regulation [4], automotive manufacturers have to certify that HIC(d) will not exceed 1000 when impacted with a 4.5 kg free motion head form with a speed of 6.7 m/s. The head form needs to be oriented in a manner so that the impact is nearly perpendicular to the target surface and thereby is likely to give a maximum HIC(d) [4, 14]. HIC is calculated as:

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

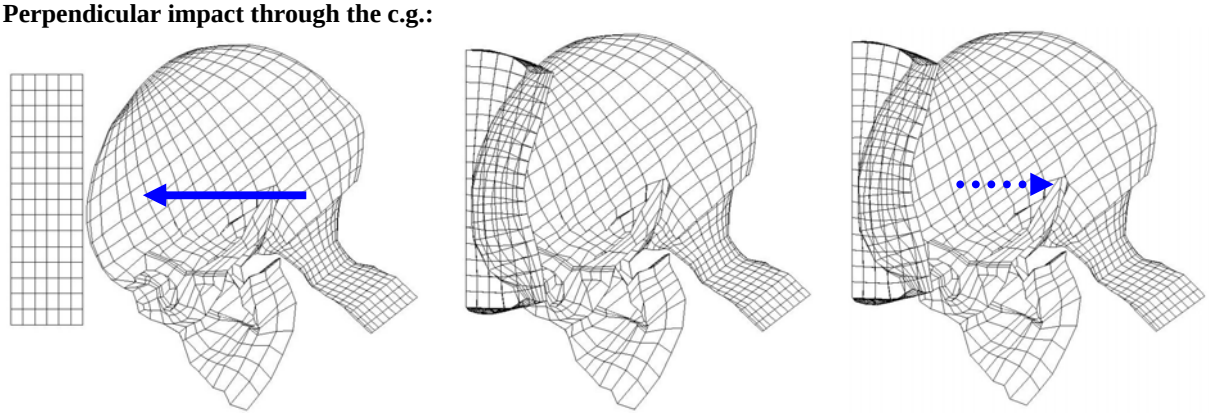
where a is the resultant head acceleration expressed as a multiple of the gravitational acceleration g, and t_1 and t_2 are any two points in time during the impact which are separated by 36 ms or less giving the maximum HIC. HIC(d) is empirically computed

using the free motion HIC to account for the neck restraint [14] in a Hybrid III dummy according to:

$$HIC(d)=0.75446 \cdot HIC+166.4 \qquad (2)$$

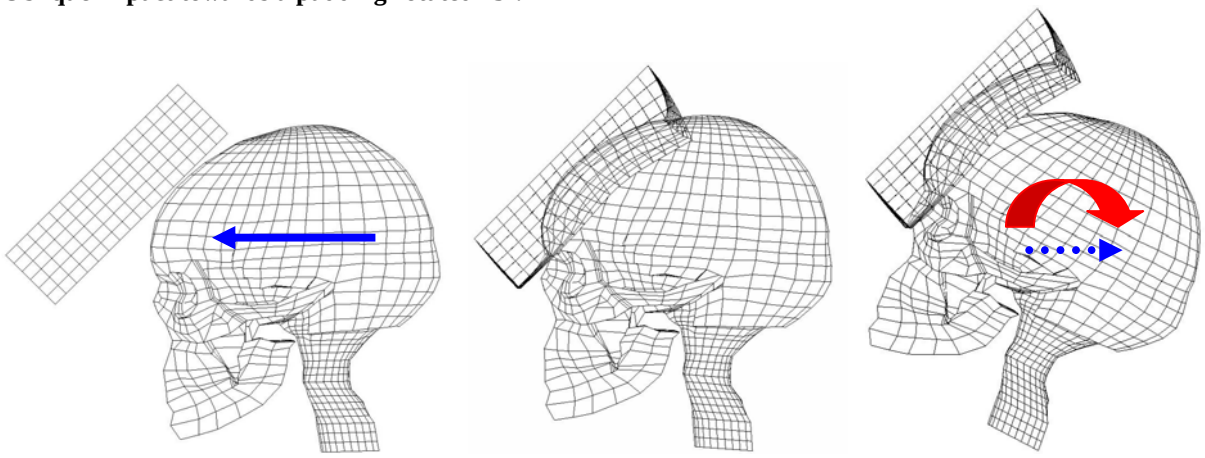
The head models are henceforth referred to as Hybrid III, and human head, respectively, were impacted towards a 50 mm thick interior padding having a 170∗170 mm contact surface with an initial velocity of 6.7 m/s (Figure 3). Perpendicular impacts through the center of gravity of the head models were simulated. Additionally, the padding was tilted 45° to the horizontal plane in an effort to evaluate the influence of an oblique impact. This was done for the choice of padding parameters giving the lowest strain the brain for the perpendicular impact case.

Perpendicular impact through the c.g.:



t=0 *t=12 ms* *t=30 ms*

Oblique impact towards a padding rotated 45°:



t=0 *t=12 ms* *t=30 ms*

Figure 3. Animation of a perpendicular impact through the center of gravity of the head model (upper) and an oblique towards a 45° tilted padding (lower).

Foam material properties

The material characteristics of expanded polypropylene (EPP) foams have recently been found to be well described (R=0.969-0.999) by a simple empirical relationship which describes the stress-strain as a function of the foam density (Equation 3) for a wide range of densities (31-145 kg/m^3) [27]. The formulation is:

$$\sigma = A(1-e^{(-E/A)\varepsilon(1-\varepsilon)^m}) + B(\frac{\varepsilon}{1-\varepsilon})^n \qquad (3)$$

where σ and ε are engineering stress and engineering strain, respectively, considered positive in compression, and A, B, E, m and n are empirical

constants derived for the particular type of foam. To create an even wider range of material behavior, the material characteristics of a theoretical EPP foam having a density of 14 kg/m^3 was generated and implemented (Figure 4). The EPP foams were modeled using a constitutive model developed for crushable foams in ls-dyna [15].

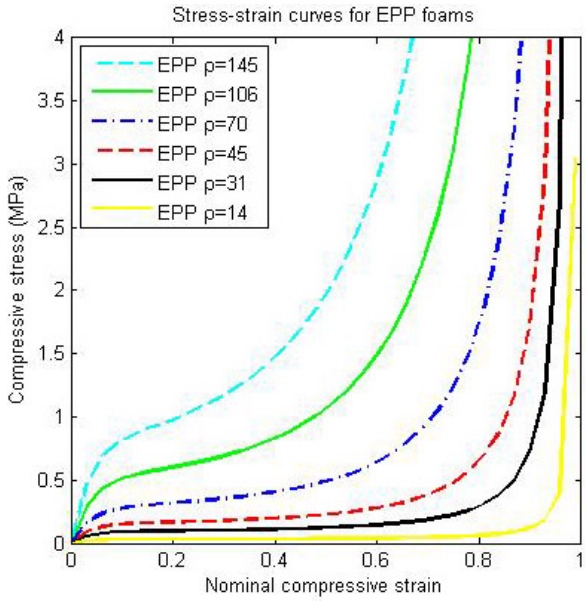


Figure 4. Material characteristics for the EPP foams used in the present study.

Interior contact definition

In order to keep the foam material elements from inverting when compressed under high pressure, an interior contact was defined. *CONTACT_INTERIOR was used in ls-dyna to account for the force transition within the foam, which is especially important when it bottoms out. It was defined so that when one layer of the foam reaches a compression strain of 98%, the internal contact transfers the loading to another layer of foam or to the scalp of the head model (or the rubber skin of the hybrid III model).

RESULTS

The resulting acceleration curves for the lowest and highest densities, as well as for one creating a low acceleration peak is seen in Figure 5. The load and acceleration curves were filtered using an SAE 1000 low-pass filter. It can be seen that the 14 kg/m^3 foam has the lowest acceleration initially until it bottoms out at a foam compression of 98%. This phenomenon is creating a short duration high spike where the load is transferred to the scalp, skull, dura, CSF and the brain (Table 2).

Different results were obtained from the parameter study with the rigid Hybrid III dummy head when compared to the human head model (Table 2, Figure 6-7). It can be seen that, despite having the same translational mass and initial velocities, the hybrid III model predicts the lowest HIC(d) value for a higher density and stiffer foam than the human head model does; The hybrid III model predicts the lowest HIC(d) value for the 45 kg/m^3 foam while the human head model predict the lowest value for the 31 kg/m^3 foam (Table 2).

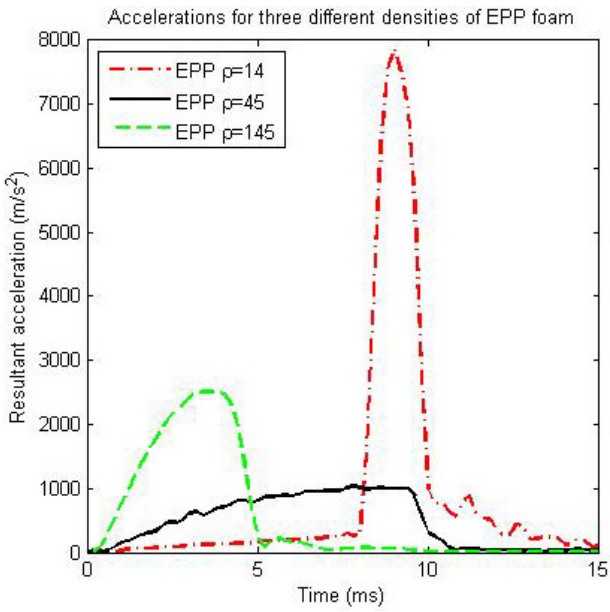


Figure 5. Curves showing the resultant acceleration of the head model using three different densities of EPP foam.

However, the HIC(d) for the head model manage to predict the strain in the brain for a purely perpendicular impact (Table 2).

Table 2.
Summary of the results from the parametric study using the rigid Hybrid III dummy head and the human head model.

	Density (kg/m^3)	14	31	45	70	106	145
Hybrid III	Peak acceleration (m/s^2)	6157	1616	1089	1398	1837	2331
	HIC(d)	6964	578	573	813	1151	1608
Head model	Peak acceleration (m/s^2)	7777	1225	1034	1397	1905	2515
	HIC(d)	11680	488	553	805	1205	1659
	Max princ strain in brain	34.8	8.2	10.0	12.6	14.3	16.3
	Max princ strain in Corp. Call.	19.2	4.0	4.1	4.5	4.7	4.9
	Max princ strain in White M.	34.1	8.2	10.0	12.6	14.3	16.3
	Max princ strain in Gray M.	34.8	6.7	8.1	10.2	11.9	13.6
	Max princ strain in Br.St.	17.4	7.4	8.9	11.2	12.5	14.3
	Max princ strain in Thal./Mid.Br.	14.5	3.7	4.0	4.6	5.1	5.4
	von M. stress in outer compact bone (MPa)	90.0	14.1	7.3	12.5	22.0	34.6
	von M. stress in inner compact bone (MPa)	90.0	13.1	10.1	14.1	20.7	27.8
	von M. stress in por. Bone (MPa)	29.3	1.5	0.7	1.1	1.9	3.1

Also, the lowest stress in the compact and porous cranial bone is found for the 45 kg/m^3 foam which correspond to the lowest values of HIC for the hybrid III head as well as the linear acceleration for the human head model (Table 2). However, the lowest strain in the brain is found for the 31 kg/m^3 foam.

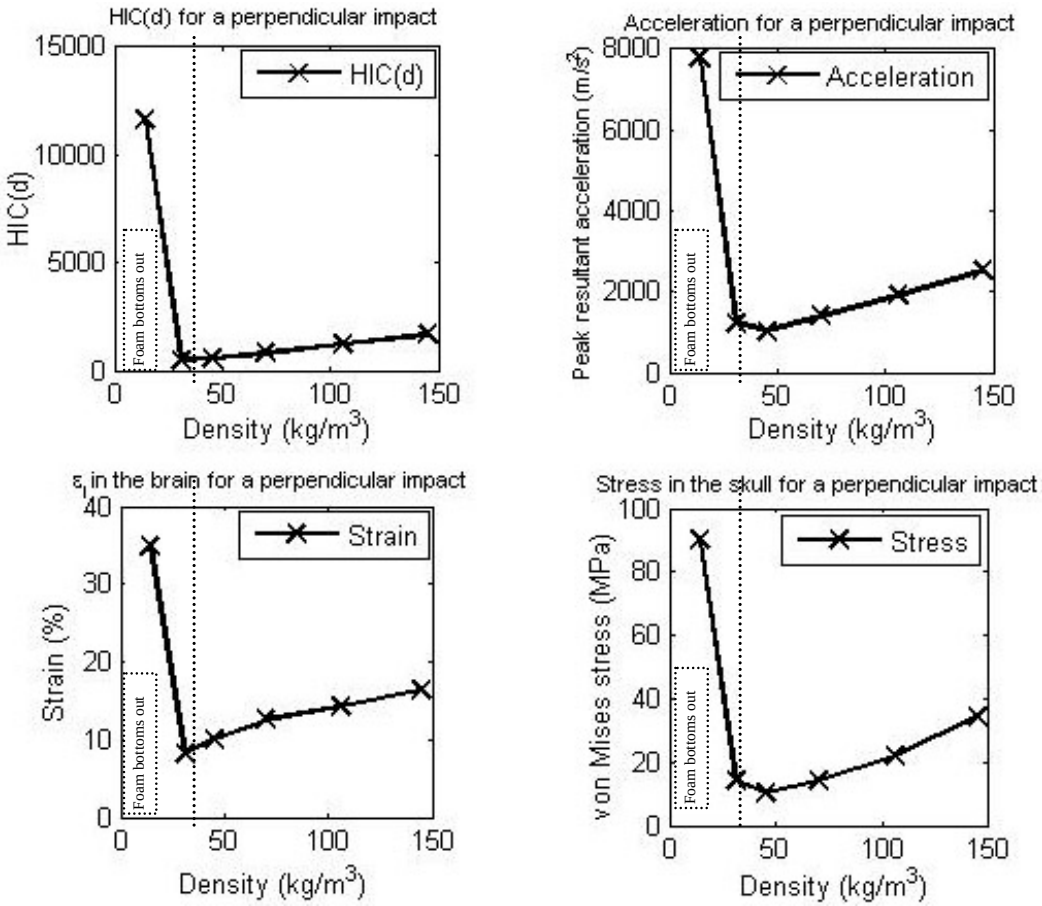


Figure 6. HIC(d) and peak resultant translational acceleration for a perpendicular impact using the biomechanical head model.

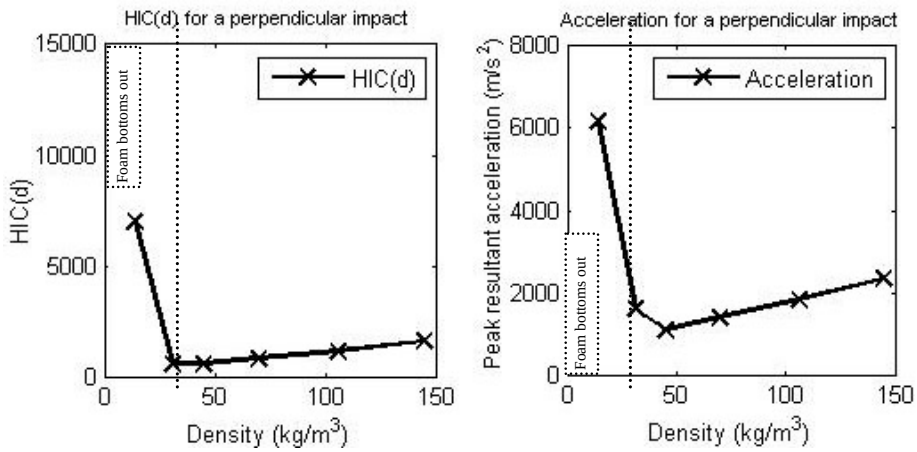


Figure 7. HIC(d) and peak resultant translational acceleration for a perpendicular impact using the HIII dummy model.

When simulating an oblique impact using the foam giving the lowest strain in the brain for the perpendicular impact (31 kg/m^3) it was found that the HIC(d) was reduced by more than 50 percent while the strain in the brain increased more than four times (Figure 8).

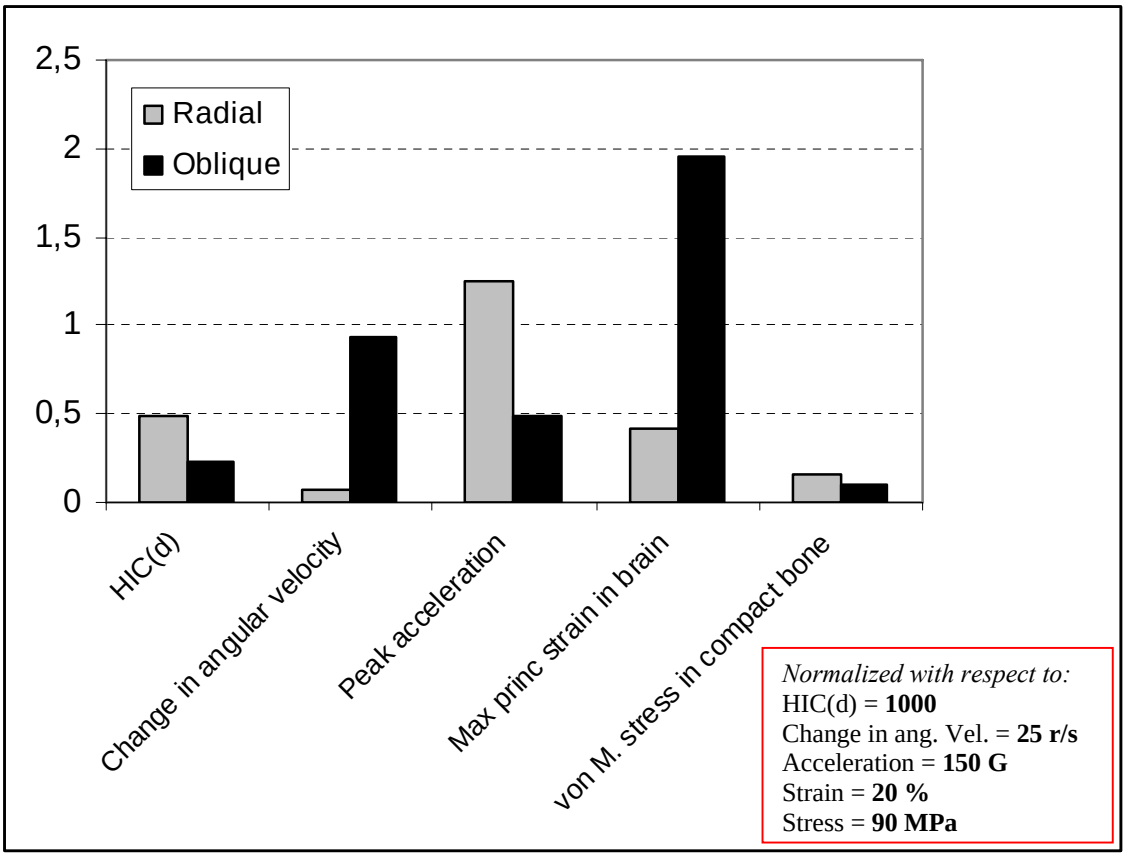


Figure 8. Staple chart summarising the HIC, strain in the brain and stress in the skull for an oblique and perpendicular impact towards the same padded surface.

It is obvious that substantially higher strain levels in the brain are obtained for an oblique impact, compared to a corresponding perpendicular one, when impacted towards the same padding using an identical initial velocity of 6.7 m/s (Figure 9).

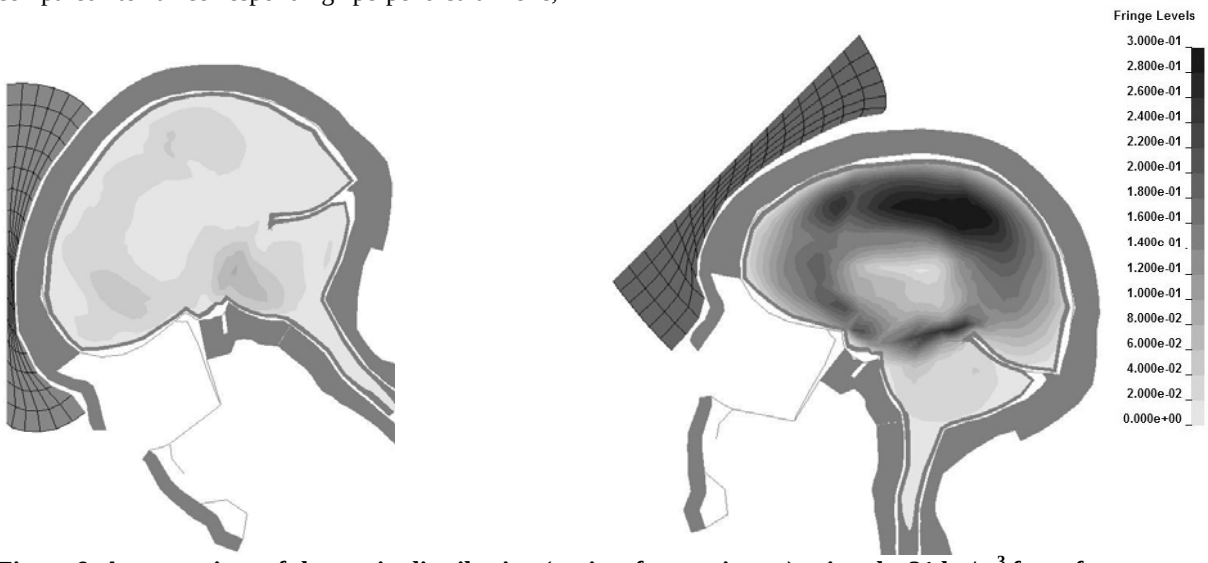


Figure 9. A comparison of the strain distribution (at time for maximum) using the 31 kg/m^3 foam for a perpendicular (left) and an oblique impact (right).

DISCUSSION

The best choice of material properties with respect to a perpendicular impact using the Hybrid III head model reached different results compared to an impact with the biomechanical head model. On the other hand, the HIC(d) for the head model manage to predict the strain level in the brain for the purely perpendicular impact. This is supporting the findings of a correlation between the probability of concussion and HIC using predominantly translational concussion data from the NFL [28]. Recently, it was also found that HIC manage to predict the strain level in the brain of an FE model for purely translational impulses of short duration [3], while the peak change in angular velocity showed the best correlation with the strain levels for purely rotational impulses.

The foam giving the lowest strain in the brain was the one with a density of 31 kg/m^3. This EPP foam has crush strength at 50 % compression of 125 kPa. The foam giving the lowest HIC(d) in the hybrid III dummy was the one with a density of 45 kg/m^3 and a stress at 50 % compression of 230 kPa. This is in correspondence with Chou *et al*. [11] who found that the crush strength for a 50 mm thick B-pillar foam pad should be lower than 345 kPa to keep the HIC below 700.

The difference in the strain in the brain between an oblique and perpendicular impact with same initial velocity towards the same padding was not predicted by the HIC(d). Even though the HIC reached acceptable levels for both the perpendicular and oblique impact towards the 31 kg/m^3 EPP padding, the maximum strain in the human brain model for the oblique impact was almost twice the suggested allowable levels [29, 30]. One of the reasons for this is that rotational effects are transferred to the head when the impact has a tangential component. These induced rotations are known to cause large shear strains in the brain tissue [31, 32]. A low HIC(d) value is predicted for the oblique impact while higher levels strains are found compared to a corresponding perpendicular impact in the same direction. This underlines findings by previous investigators [32] who subjected 25 squirrel monkeys to controlled sagittal plane head motions, and found greater frequency and severity of brain lesions after rotation. This is consistent with the results presented herein, as well as the hypothesis presented by Holbourn [31].

For the pure perpendicular impact an insignificant peak change in angular velocity is found together with relatively low strain levels in the brain. For the oblique impact a large strain level is found in the brain for a large peak change in angular velocity. This corresponds to Holbourn's hypothesis [31] that the strain (and the injury) is proportional to the change in angular velocity for rotational impulses of short durations. Margulies and Thibault [33] presented a criterion for DAI described as tolerance curves of angular accelerations as a function of peak

change in angular velocity. Judging from those curves, angular accelerations exceeding ca. 8 krad/s^2 combined with an angular velocity of 70 rad/s or higher gives a risk of injury in the adult [33]. For the oblique impact in the present study, an angular acceleration of 3.3 krad/s^2 and a peak change in angular velocity of 23.5 rad/s was found together with a maximum strain in the brain of 39 percent which is almost twice the suggested tissue level tolerances for DAI [29, 30]. On the other hand, the HIC(d) is not insignificant for the oblique impact. Probably, a combination of the peak change in angular velocity and HIC(d) would predict the difference between perpendicular and oblique impacts of various severities. In this study, impact at only one location of the head is studied and therefore the results might differ depending on what impact location that is chosen. However, an impact to the forehead region was chosen in this study and this region is known to withstand more violence than most other parts of the head both for DAI [32] and for skull fractures [34]. Therefore the presented stress and strain levels for the head would probably be even higher if the impact was from other directions.

Strich [35] found diffuse degeneration of white matter in the cerebral hemispheres, as well as in the brain stem and corpus callosum areas in patients who have endured severe head trauma. This indicates that high strain in the white matter adjacent to the cortex, as seen in Figure 9 of this study, is likely to occur in a real life accident. Correspondingly, low levels of strain can be seen in the vicinity of the ventricles in the model, which supports the hypothesis that a strain relief is present around the ventricles [36].

The bulk modulus of brain tissue [37] is roughly 10^5 times larger than the shear modulus. Thus, the brain tissue can be considered as a fluid in the sense that its primary mode of deformation is shear. Therefore, distortional strain was used as an indicator of the risk of traumatic brain injury. The maximal principal strain was chosen as a predictor of CNS injuries since it has shown to correlate with diffuse axonal injuries [29, 30, 38, 39, 40, 41], as well as for mechanical injury to the blood-brain barrier [42]. Other local tissue injury measures have also been proposed and evaluated, such as von Mises stress [42, 43, 44], the product of strain and strain rate [45, 46, 47], the strain energy [42], and the accumulative volume of brain tissue enduring a specific level of strain, the Cumulative Strain Damage Measure (CSDM), [9, 48]. However, a correlation has recently been found between the brain injury pattern of a patient being the victim of a motocross accident and the strain pattern in the head model [49]. This strain is very sensitive to the choice of stiffness for the brain tissue [17] and more work is needed to fully describe the non-linear and viscoelastic response of living brain tissue.

Another possible limitation is the constitutive model used for the foam. However, this model has shown to predict the response in uniaxial compression tests for expanded polystyrene foams of similar densities in a previous study [50]. On the other hand, to model the elastic spring-back of low density foams such as polyurethane foams, probably a different constitutive model should be chosen. Also, the high load and acceleration behavior created when the foam bottoms out is sensitive to the parameters chosen for the foam and the interior contact. In this study, the stress-strain curves were defined up to 99 % compression for all foams, while the interior contact was activated when the strain reached 98 %.

In the extension, protective devices and materials can be optimized to see if the tissue level stresses and strains can be minimized so that the potential consequence in a future accident could be reduced or avoided. Iterative optimization procedures in conjunction with dynamic and non-linear FEA can be used together with detailed FE models to maximize the safety for the humans when impacting towards of interior and exterior surfaces in automotive structures. The existing FE model of the head can be used in optimization of the properties and geometry of energy absorbing materials, so that the stresses and strains on a tissue level are minimized. This methodology has previously been used for optimization of simplified hood structures [51]. The proposed methodology is directly applicable in development of interior and exterior surfaces in heavy vehicles and rail vehicles as well.

CONCLUSIONS

The results emphasize the importance of treating the human brain as a non-rigid body. Although it is obvious it must be kept in mind that in strive for improved safety it is essential to employ physically representative metrics since the applied criteria will drive the development. Hence, local tissue thresholds or more human-like dummies together with injury criteria accounting for both angular and translational kinematics should be used to obtain more physically representative and reliable optima in safety design. This result is conceptually obvious since a global criterion will never cover all the various injury mechanisms characterized by local tissue deformation.

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