INFLUENCE OF DIRECTION AND DURATION OF IMPACTS TO THE HUMAN HEAD EVALUATED USING THE FINITE ELEMENT METHOD

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ABSTRACT

The objective of the present study was to analyze the effect of different load directions and durations following impact using a detailed finite element model of the human head. It was found that the influence of impact direction had a substantial effect on the intracranial response. When evaluating the global kinematic injury measures for the rotational pulses, the change in angular velocity corresponded best with the intracranial strains found in the FE model. For the translational impulse, on the other hand, the HIC and the HIP showed the best correlation with the strain levels found in the model.

Key words: Brain, Finite Element Method (FEM), Head Injury Criterion (HIC), Head Impact Power (HIP).

IN EUROPÉ, ROAD ACCIDENTS are the second most frequent cause of death preceded only by cancer (European Transport Safety Council, 1999). The total annual rate of head injuries in Sweden over the last 14 years is relatively constant (Kleiven *et al.*, 2003). Thus, in spite of several national preventive strategies, there has not been an important impact on the total burden of head injury. For people younger than 45 years, the frequency of death or severe injury from road accidents is about six times higher than that from cancer. A significant number of road accidents affect the Central Nervous System (CNS) in a devastating way by transferring high kinetic energy to the nervous tissue. Subdural hematomas (SDH) and diffuse axonal injuries (DAI) are more lethal than most other brain lesions (Gennarelli and Thibault, 1982). This gives a special interest in deriving injury criteria for SDH and DAI. Gennarelli (1983) suggested that SDH was produced by short duration and high amplitude of angular accelerations, while DAI was produced by longer duration and low amplitude of coronal accelerations. A threshold for DAI has been proposed (Margulies and Thibault, 1992) which accounts for rotational impulses in the coronal plane. Moreover, studies by Ueno and Melvin (1995), and DiMasi *et al.* (1995) found that the use of either translation or rotation alone may underestimate the severity of an injury. Generally, the head injury criterion (HIC) (National Highway Traffic Safety Adm., 1972) is used when evaluating the consequences of an impact to the head. HIC is based exclusively on the resultant translational acceleration of the head. Thus, HIC and proposed acceleration thresholds do neither take into consideration rotational and translational loads, nor directional dependency. There is therefore a need for more complex injury assessment functions, accounting for both translational and angular acceleration components as well as changes in the direction of the loading.

When a comparison between translation and rotation has been performed, the usual approach has been to compare a non-centroidal rotational impulse with a translational impulse giving a similar acceleration measured at the center of gravity (c.g.) (Margulies *et al.*, 1985, Bandak and Eppinger, 1994). This gives a good basis for criticism of head injury criteria based solely on the translational acceleration (i.e. HIC). In this case, however, the comparison will be between a translational impulse and an equal translational impulse in addition to the induced rotational one. A more objective approach could be to apply the same dosage of mechanical energy per time unit (i.e. the power) for the separate degrees of freedom as described here, and proposed as a new head injury criterion: HIP (Newman *et al.*, 2000).

The influence of certain impact directions have been investigated for DAI (Gennarelli *et al.*, 1982, 1987) and cerebral concussion (Hodgson *et al.*, 1983). In both studies, subhuman primates were used. In a three-dimensional (3D) numerical study (Zhang *et al.*, 2001), brain responses between frontal and lateral impacts were compared. That study confirmed earlier results by Gennarelli *et al.* (1982) that loads in the lateral direction are more likely to cause DAI than impulses in the sagittal plane. Zhou *et al.* (1995) suggested that SDH is more easily produced in an occipital impact than in a corresponding frontal one. Later, the same researchers (Zhou *et al.,* 1996) found that AP motion causes higher strain in the bridging veins than a corresponding lateral motion. However, in all these numerical studies, a tied interface was imposed

between the skull and the brain leaving out any possibility of evaluating relative motion induced injuries such as SDH. Recently it was found that the influence of impact direction had a substantial effect in the prediction of subdural hematoma (Kleiven, 2003).

Due to the limited studies of impact directions, the existing head injury criteria could not be evaluated for all types of impacts. Recently, a new global kinematic-based head injury criterion, called the HIP was presented (Newman *et al.*, 2000). In that study, it was proposed that coefficients for the different directions could be chosen to normalize the HIP with respect to some selected failure levels for a specific direction. However, values of the coefficients were not presented and information regarding directional sensitivity was lacking.

Another important issue in modeling of the human head is the selection of material properties for various intracranial structures. The three-dimensional (3D) models use linearly elastic or viscoelastic constitutive properties and conventional (displacement-based) finite element formulations that can create severe numerical instabilities when dealing with nearly incompressible materials. The choice of shear properties for the brain tissue is difficult since the span of published values varies several orders of magnitude. Donnelly (1998) reviewed and reported the average values of the shear relaxation modulus for brain tissue. According to that study, the average value of the instantaneous shear relaxation modulus for brain tissue is the order of 1 kPa. Most 3D FE modeling studies have included properties that are around 10-1000 times larger than the average published values.

Thus, the aim of the present investigation was to study the influence of inertial forces on all the degrees of freedom of the human head, evaluated with a detailed FE model. Global kinematic measures such as magnitude in angular acceleration, change in angular and translational velocity, HIC, as well as HIP, were investigated with regard to their ability to take into account consequences of different impact directions and durations for the prediction of intracranial strains associated with injury.

METHODS

FINITE ELEMENT MESH

A detailed and parameterized Finite Element (FE) model of the adult human head was created, comprising the scalp, skull, brain, meninges, cerebrospinal fluid (CSF), and eleven pairs of parasagittal bridging veins (Fig.1). A simplified neck, including an extension of the brain stem into the spinal cord, the dura and pia mater, the vertebrae and muscles, was also modeled.

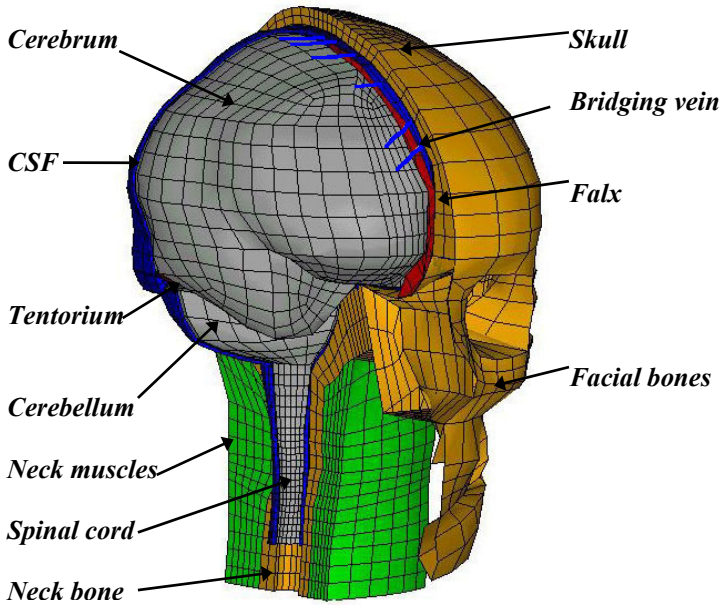

Fig. 1 – Finite element mesh of the human head.

This model has been experimentally validated against pressure data in a previous study (Kleiven and von Holst, 2002a) as well as relative motion magnitude data (Kleiven and von Holst, 2002b). Also, a comprehensive correlation between the FE model output and the relative motion between human cadaver

brain and skull in anatomical X, Y, and Z components has been demonstrated for three impact directions (Kleiven and Hardy, 2002). The model has been validated with experiments performed using acceleration impulses of magnitudes and durations close to the ones in the present study.

MATERIAL PROPERTIES

To cope with the large elastic deformations, a Mooney-Rivlin hyperelastic constitutive law was utilized for the CNS tissues. Mendis *et al.* (1995) derived the rate dependent Mooney-Rivlin constants C_{10} and C_{01} and time decay constants β_i, using experiments published by Estes and McElhaney (1970) on white matter from the corona radiata region.

According to Kleiven and Hardy (2002), the average brain stiffness properties reported by Donnelli (1998) showed the best correlation with experiments on localized motion of the brain. Previous studies have, on the other hand (Metz *et al.*, 1970; Stalnaker *et al.*, 1977), indicated that the stiffness of the brain tissue might decrease after death. Therefore Mooney-Rivlin constants corresponding to an effective (long-term) shear modulus of around 520 Pa was used for most of the analysis. Since the strain is highly sensitive to the shear modulus (Kleiven and Hardy, 2002), three additional properties corresponding to effective long-term shear modulii of 130-2600 Pa were also applied to further investigate the sensitivity of brain stiffness properties due to a rotational and a translational motion. The stiffness parameters C_{10}, C_{01}, G_1, and G_2 were scaled while the decay constants were not altered. The law was introduced for the white matter and the gray matter, which is reported to be insignificantly (about 4 %) stiffer than white (Prange *et al.*, 2000). The Mooney-Rivlin constants for the brain stem were assumed to be 80 % higher than those for the gray matter in the cortex (Arbogast and Margulies, 1997). For the spinal cord and cerebellum, the same properties as for the white and gray matter were assumed due to lack of published data. A summary of the properties for the other tissues of the human head used in this study is shown in table 1 below.

Table 1 – Properties used in the numerical study.

Tissue	**Young's modulus** [MPa]	**Density** [kg/dm^3]	**Poisson's ratio**
Outer table/Face	15 000	2.00	0.22
Inner table	15 000	2.00	0.22
Diploe	1000	1.30	0.24
Neck bone	1000	1.30	0.24
Neck muscles	0.1	1.13	0.45
Brain	*Hyperelastic/Viscoelastic*	1.04	0.4999994-0.49999997
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
Pia mater	11.5	1.13	0.45
Scalp	16.7	1.13	0.42
Bridging veins	*EA=1.9 N*		

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

INTERFACE CONDITIONS

The dura is often adhered to the skull, thus the interface between the skull and the dura was modeled with a tied contact definition in LS-DYNA (Livermore Software Technology Corporation, 2001). Because of the presence of CSF between the meningeal membranes and the brain, sliding contact definitions were used for these interfaces. The chosen contact definition allowed sliding in the tangential direction and transfer of tension and compression in the radial direction. This was done in part because a fluid structure interface is likely to experience a vacuum when a pressure wave reflects at the contrecoup site, or when inertia forces create tension in brain regions opposite to the impact. An average CSF thickness of roughly 2 mm was used, which corresponds to approximately 120 ml of subdural and subarachnoidal CSF. A coefficient of friction of 0.2 was used, as proposed by Miller *et al.,* (1998). The subdural and subarachnoidal CSF, as well as the ventricular CSF, was modeled with eight node brick elements and a fluid element formulation. The outer boundary of the elements representing the ventricles was joined to the brain tissue elements through common nodes.

APPLIED LOADS FOR STUDY OF DIRECTIONAL INFLUENCE

A total of nine acceleration pulses (pure translation and angular) were applied to the center of gravity of the head in the ± PA, ± SI, and in the lateral directions (Fig. 2), in order to look into directional differences and to derive the scaling factors in Eq. (7). In the study of the angular acceleration components, a squared sinusoidal pulse *(sin^2)* with an amplitude of 10-11.6 krad/s^2 and a duration of 5 ms, resulting in a peak angular velocity of 25-29 rad/s (in the range of the proposed threshold for DAI by Margulies and Thibault, 1992), giving a HIP_{max} of 4.3 kW for all directions. To obtain a comparison with the angular impulses, a squared sinusoidal pulse with an amplitude of 794 m/s^2 (80 g) and a duration of 5 ms was used for the translational impulses resulting in a HIC of 52 and a HIP_{max} of 4.3 kW.

EVALUATION OF HIP, α, AND Δω AS AN INJURY PREDICTOR FOR ROTATIONAL KINEMATICS

In addition to the directional study, the various global kinematic-based injury measures were evaluated using the same impulse shapes by keeping the measures constant and varying the impulse duration as seen below. If the measure is correlating with strain (which is supposed to correlate with injury), applying a constant value of the injury measure (while varying the duration) would result in a constant strain in the model.

For the rotational kinematics, the peak angular acceleration, change in angular velocity, and HIP were evaluated. First, a constant AP angular acceleration impulse with an amplitude of 10 krad/s^2 was used. The duration was varied leading to a HIP_{max} of 1.08-17.3 kW and a change in angular velocity, Δω, of 6.25-50 rad/s. In addition, a constant change in angular velocity, Δω, of 25 rad/s was applied while the duration was varied leading to a HIP_{max} of 1.08-8.7 kW and a peak angular acceleration, α, of 2.5-20 krad/s^2. Finally, HIP was evaluated by keeping a constant value of 4.3 kW, while the duration was varied leading to a change in angular velocity, Δω, of 17.675-50 rad/s and a peak angular acceleration, α, of 5.0-14.14 krad/s^2.

EVALUATION OF HIC, HIP AND ΔV AS AN INJURY PREDICTOR FOR TRANSLATIONAL KINEMATICS

For the translational AP direction, the HIC was evaluated by keeping a constant value of 1000 while the duration was varied leading to a change in velocity, ΔV, of 4.27-14.87 m/s and a peak acceleration of 1487-3417 m/s^2. In addition, the HIP and change in velocity were kept at a constant level of HIP_{max}=46 kW and ΔV=6.47 m/s, respectively, for the various impulse durations.

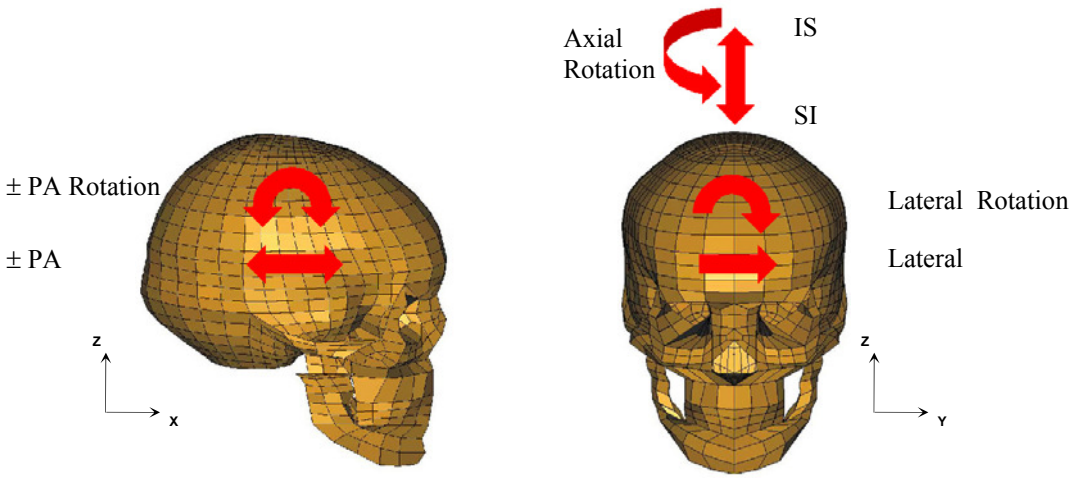

Fig. 2 – Load directions for translational and angular acceleration pulses.

The models were used to investigate the differences in terms of maximal principal strain in the brain due to variation in impact direction and duration. Furthermore, the Head Injury Criterion (HIC), the recently proposed Head Impact Power (HIP) criterion, as well as peak angular acceleration and change in angular velocity were evaluated with respect to the strain in the central nervous system (CNS) tissue. Thus, pulses of the same shape were applied to evaluate:

1. The sensitivity of impact direction by applying impulses resulting in constant values of HIP and HIC as previously described in Kleiven (2003).

2. The proposed global kinematic-based injury measures (HIC, HIP, peak angular acceleration and change in angular/translational velocity) by varying the duration and keeping the measure constant.
3. The sensitivity of brain shear stiffness on the intracranial strain.

The maximal principal strain was chosen as a predictor of CNS injuries since it has been proposed as a predictor of diffuse axonal injuries (DAI), (Bain and Meaney, 2000).

REVIEW OF HEAD INJURY CRITERIA

Head Injury Criteria (HIC)

The most common criterion used to predict head injuries is *the Head Injury Criterion (HIC)* (1972, National Highway Traffic Safety Adm.). Researchers used an indirect approach to study human concussion by impacting embalmed cadaver heads, and look for skull fractures. The rationale for using this indirect approach was based on the clinical observation that concussion is present in 80 percent of patients with simple linear skull fractures. It is based on the resultant translational acceleration of the head. The HIC should not exceed 1000 if the integration limits are separated no more than 36 ms. The acceleration, a, in this formula is given in g.

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

The basis underlying HIC was first introduced as a curve fit to the Wayne State Tolerance Curve (WSTC). The WSTC (Fig. 3) was first presented by Lissner (1960), and was generated by dropping embalmed cadavers onto unyielding, flat surfases, striking the subject on the forehead. Gadd (1966) developed the Severity Index (SI) to fit the WSTC (Fig. 3), with a value greater than 1000 considered to be dangerous to life. It was not only based on the WSTC, but also upon additional long pulse duration data by means of the Eiband (1959) tolerance data and other primate sled test data. The SI provided a good fit for both the short duration skull fracture data and the longer duration Eiband data out to 50 ms duration.

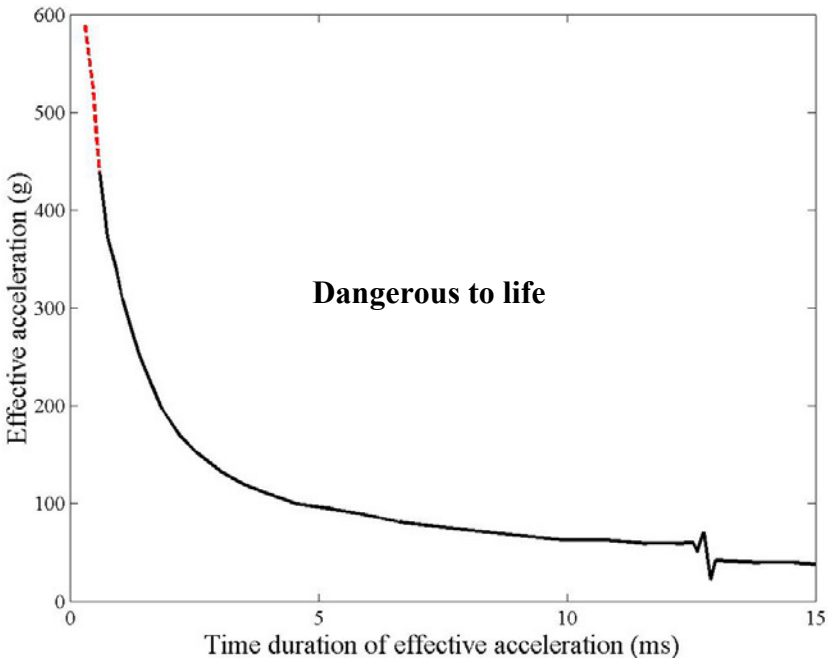

Fig. 3 – The Wayne State Tolerance Curve (Redrawn from Versace, 1971).

Head Impact Power (HIP)

Recently a new global head kinematic-based injury potential measure, called the Head Impact Power (HIP) was presented (Newman *et al.*, 2000). In that study, it was proposed that coefficients for the different directions could be chosen to normalize the HIP with respect to some selected failure levels for a specific

direction. However, values of the coefficients were not presented and information regarding directional sensitivity was lacking.

$$HIP = ma_x \int a_x dt + ma_y \int a_y dt + ma_z \int a_z dt + I_{xx}\alpha_x \int \alpha_x dt + I_{yy}\alpha_y \int \alpha_y dt + I_{zz}\alpha_z \int \alpha_z dt \qquad (2)$$

The x-axis was defined along the Posterior-Anterior (PA) direction, the y-axis along the lateral-direction, and the z-axis in the Inferior-Superior (IS) direction.Head injury is assumed to correlate with the maximum value of HIP achieved by equation 2 during an impact, named HIP_{max}.

Generalized Acceleration Model for Brain Injury Threshold (GAMBIT)

This was an early effort to combine thresholds for translational and rotational kinematics (Newman, 1986). The GAMBIT requires to establish the maximum value, G, of the function *G(t),* i.e. *G=max[G(t)]*. *G*=1 is normally set to correspond to a 50% probability of AIS>3. Some versions of *G(t)* have been presented (Newman, 1986, 2000), but the most general one is:

$$G(t) = [(\frac{a(t)}{a_c})^n + (\frac{\alpha(t)}{\alpha_c})^m]^{1/s} \qquad (3)$$

where a(t) and α(t) are the instantaneous values of the translational and rotational acceleration, respectively, and n, m, and s are empirical constants selected to fit available data. The a_c and α_c are the acceleration thresholds for a pure translational, and a pure rotational impulse, respectively. Proposed values for the constants are: n=m=s=1, a_c=250g and α_c=10000r/s^2 (Newman, 1986), and n=m=s=2, a_c=250g and α_c=25000r/s^2 (Newman *et al.*, 2000). Since no dependency of the impulse duration is included, the GAMBIT can be seen as a peak-acceleration criterion for a combined rotational and translational impulse.

ANGULAR THRESHOLDS

Proposed Threshold for DAI

Margulies and Thibault (1992) presented a criterion for DAI. It is developed using experiments on primates in combination with gel physical models and analytical scaling procedures using a cylindrical approximation. The criterion is represented by curves representing equal strain in the analytical model as a function of the angular acceleration and peak change of angular velocity. Judging from Fig. 4 (left), rotational accelerations exceeding 10 krad/s^2 combined with an angular velocity of 100 rad/s or higher gives a risk of DAI in the adult. These curves show that for small changes in angular velocities the injury is less dependent on the peak angular acceleration, while for high values of peak change in angular velocity, the injury is sensitive to the peak angular acceleration. This is in agreement with the hypothesis of Holbourn (1943). He stated that the shear strain, and thus injury, for long duration impulses (large peak change in rotational velocity) is proportional to the acceleration, while the injury is proportional to the change of velocity of the head for short duration impacts.

Proposed Thresholds for Concussion

Ommaya *et al.* (1967), and Ommaya and Hirsch (1971) proposed limits for angular acceleration (α). A more than 99% probability of concussion was estimated for α>7500 r/s^2, when the impulse duration exceeds 6.5ms (Ommaya *et al.*, 1967). Also, a limit of α>1800 r/s^2 to produce concussions due to head rotation induced by whiplash was proposed (Ommaya and Hirsch 1971).

Proposed Threshold for SDH

In a primate study (Gennarelli and Thibault, 1982), it was proposed that an angular acceleration exceeding 175 krad/s^2 combined with an impulse time exceeding 5ms, would produce SDH in the rhesus monkey (Fig. 4, right).

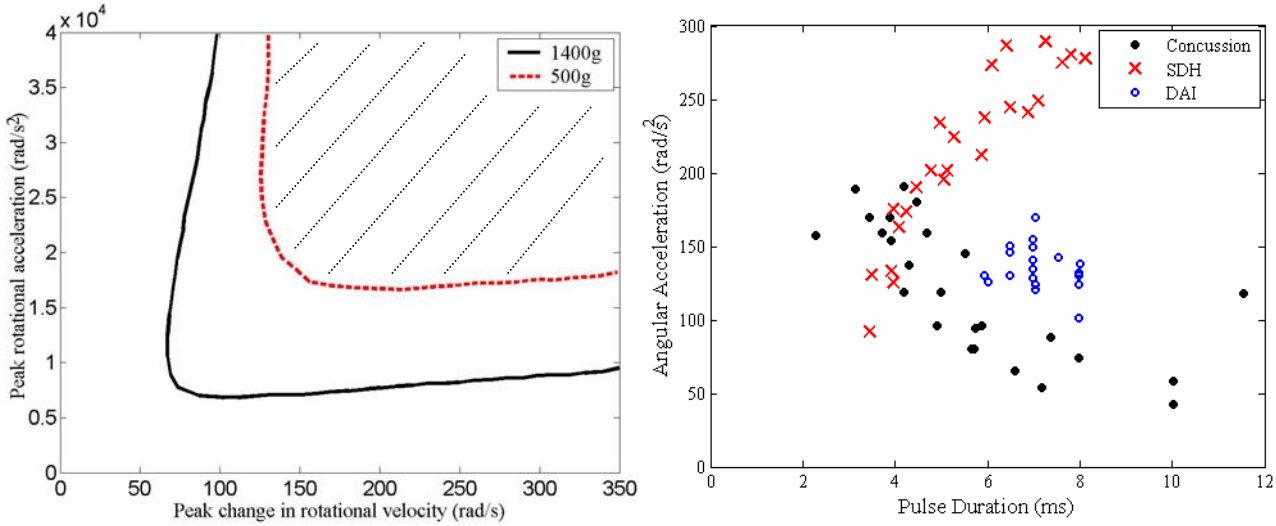

Fig. 4 – Proposed tolerance curve for DAI (left). Modified from Margulies and Thibault (1992). The full line represents the limit for an average adult (brain mass 1400g). Results from experiments on primates (right). Redrawn from results presented in Gennarelli and Thibault (1982).

Löwenhielm (1974) stated that bridging vein disruption due to rotational movement of the head is obtained when the angular acceleration exceeds 4.5 krad/s^2 and/or the change in angular velocity exceeds 50 rad/s using collision tests with cadavers. The estimation of the rotational accelerations/velocities were based on differentiation of smoothed cubic spline interpolations (of head rotations) of high speed videos (500 frames/s) of the planar motion of the head in the sagittal plane. Thus leaving out any synergic effects of other rotations/translations. On the other hand, the motions were not pure rotational, and either none or several bridging veins were ruptured indicating that a real threshold never was found. Also, in these experiments which were previously presented by Voigt and Lange (1971), there were a high level of violence other than the rotational. The non-belted cadavers were seated on a sled and accelerated to velocities of between 43 to 60 km/h before braked into standstill and impacting towards the instrument panels. In some of these experiments the translational acceleration on the top of the head was recorded. In the more severe cases, translational accelerations varying between +/- 200g was recorded, adding to the rotational violence.

RESULTS

DIRECTIONAL SENSITIVITY

A summary of the results from the comparison of translational and angular impulses in different directions is shown in Fig. 5. It can be seen that the largest strain in the brain appears for the lateral and axial rotational impulses, while substantially smaller strain is found for the translational impulses (Fig. 5).

For the angular impulses, the same HIP_{max} values are calculated as for the translational impulses, while the HIC is equal to zero for a pure rotational impulse. Nevertheless, larger strains in the brain appear for the rotational impulses. For this type of loading, the worst case is the lateral rotation where the highest strain in the cortex, corpus callosum and brain stem appears. Almost a tenfold increase in the intracranial strains is found for the PA and AP impulses, when switching from a translational to a rotational mode of motion. For the lateral direction, a smaller sensitivity to the mode of motion is found. In this case, the strain in the various areas of the brain increases about ten times when changing from a lateral translational to a lateral rotational motion.

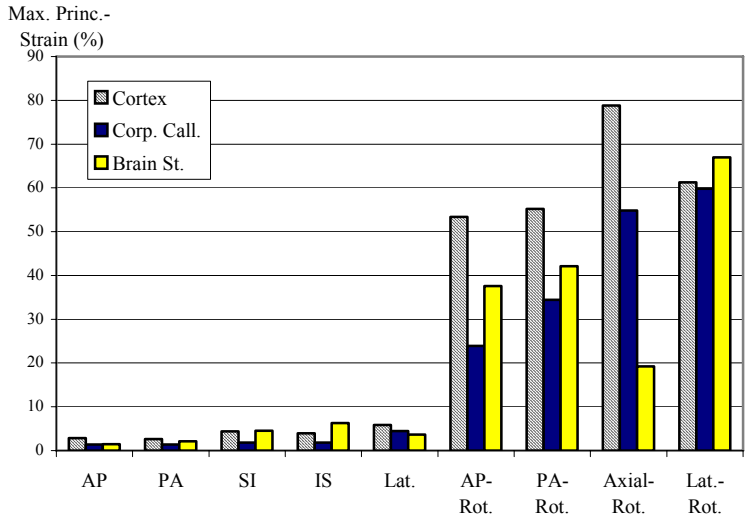

Fig. 5 – Results for different directions and translational and angular acceleration impulses. HIP_{max}=4.3 kW, HIC=0 for the angular impulses, while HIP_{max}=4.3 kW and HIC=52 for the translational impulses.

Images showing a parasagittal view of the straining of the brain when enduring the AP rotational and AP translational impulses simulating a frontal impact can be seen in Fig. 6. Note the high levels of strain close to the vertex of the skull as well as close to the irregularities in skull base for the rotational impulse. Correspondingly, low levels of strain can be seen in the vicinity of the ventricles. A pure AP rotation is not likely to occur in real life, but can most closely be compared with an uppercut in boxing, while large AP translational accelerations can be experienced during a frontal collision.

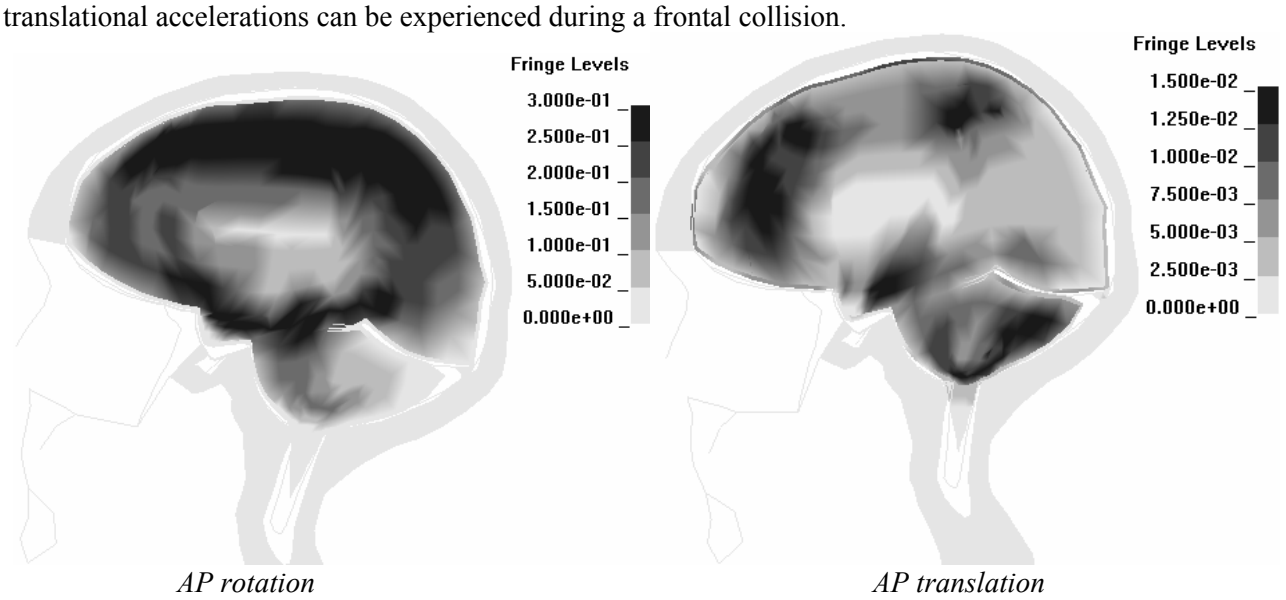

Fig. 6 – Strain distribution (around maximum) for AP rotation (left) and AP translation (right) using the same duration, impulse shape, giving the same HIP, and resulting in substantially different intracranial strains.

Figure 7 shows the strain distribution in a mid-coronal cross-section for the lateral rotational impulse (Fig. 7, upper left) and a sagittal view of a inferior-superior (IS) translational motion (Fig.7, lower right). Note the high levels of strain in the corpus callosum area, and close to the brain stem for the lateral rotation. For the IS impulse, the highest strains can be noted in the spinal cord as well as around and close to the brain stem and cerebellum. An impulse with a high level of lateral rotation could occur in a side impact during an automotive or pedestrian accident, while a pure IS translational motion can be compared with a fall accident or a helicopter crash landing.

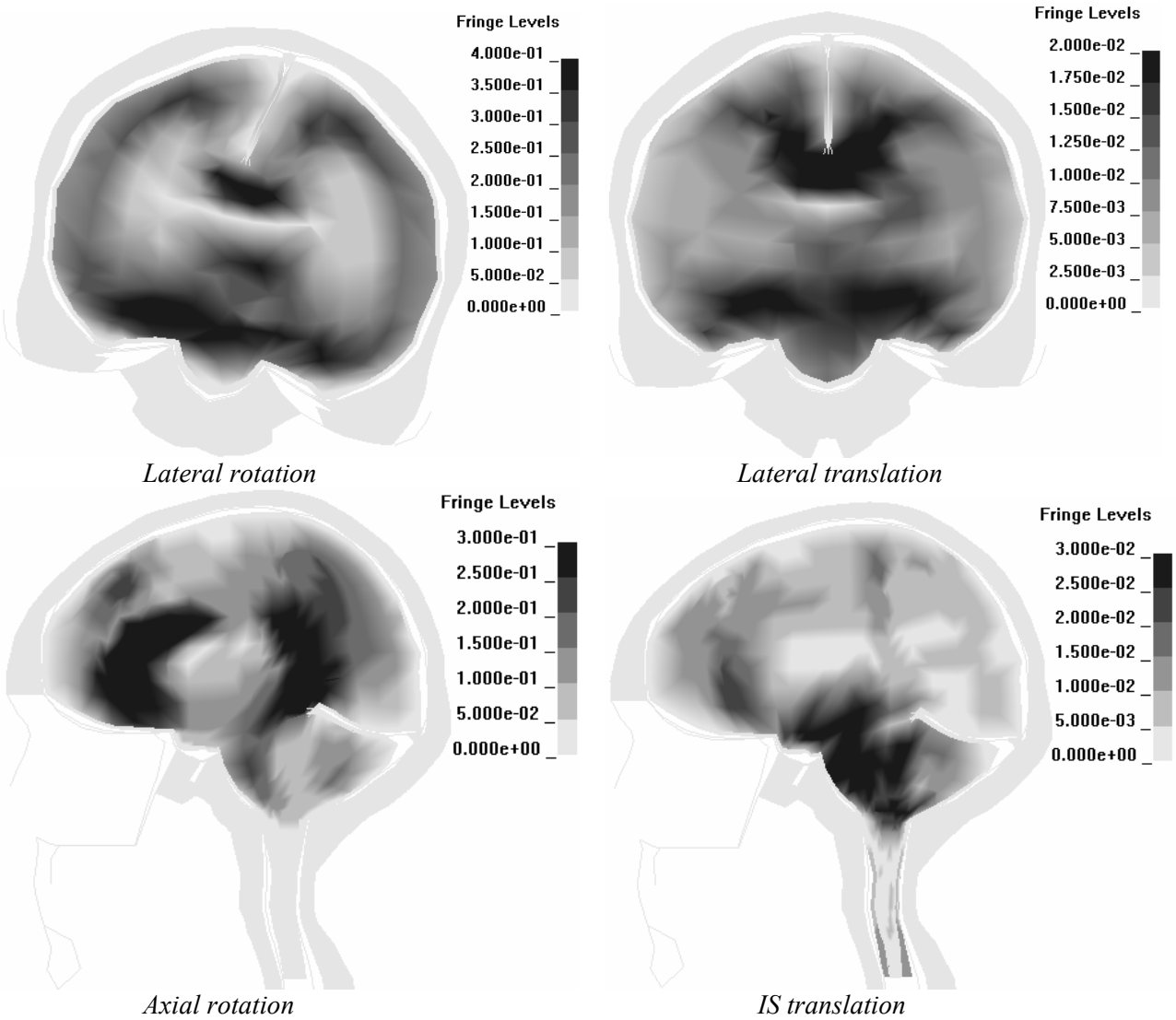

Fig. 7 – Strain distribution (around maximum) for lateral rotation (upper left), lateral translation (upper right), axial rotation (lower left) and IS translation (lower right) using the same duration and impulse shape, resulting in the same HIP.

EVALUATION OF HIP, α, AND $\Delta\omega$ AS AN INJURY PREDICTOR FOR ROTATIONAL KINEMATICS

When evaluating the various global kinematic-based injury measures for an AP rotational motion by keeping the various measure constant while varying the impulse duration, it was found that the change in angular velocity mirrored the level of strain in the brain better than the HIP and the peak angular acceleration did. An almost constant level of strain was found for a constant change in angular velocity, while for both the HIP and the peak angular acceleration gave an increasing strain level for an increase in the impulse duration (Fig. 8).

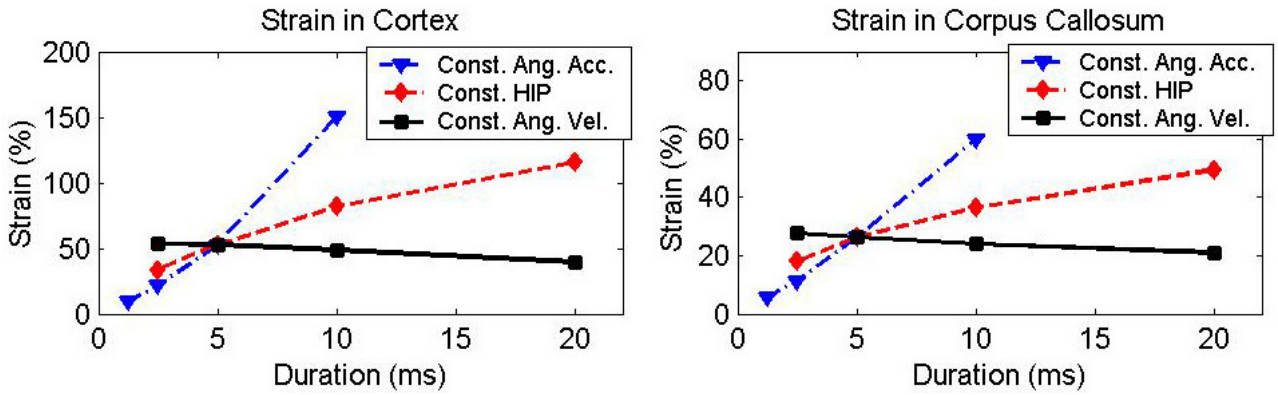

Fig. 8 – Evaluation of global kinematic measures for rotational motion; Keeping the magnitudes of angular acceleration, change in angular velocity, and the HIP, respectively, constant while varying the impulse duration.

EVALUATION OF HIC, HIP AND ΔV AS AN INJURY PREDICTOR FOR TRANSLATIONAL KINEMATICS

When evaluating the various global kinematic-based injury measures for an AP translational motion by keeping the various measure constant while varying the impulse duration, it was found that the HIC and HIP mirrored the level of strain in the brain better than the change in velocity did. An almost constant level of strain was found for a constant HIC and HIP, while a constant change in velocity gave a decreasing strain level for an increase in impulse duration (Fig. 9).

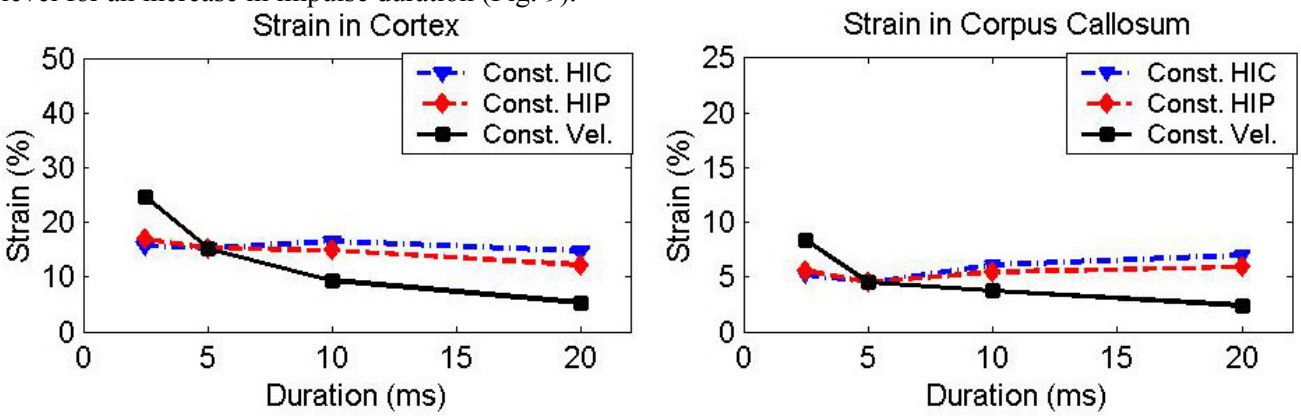

Fig. 9 – Evaluation of global kinematic measures for translational motion; Keeping the magnitudes of HIC, HIP, and change in velocity, respectively, constant while varying the impulse duration.

SENSITIVITY TO THE SHEAR STIFFNESS OF BRAIN TISSUE

Also, the strain in the brain shows a large sensitivity to the shear properties utilized for the brain tissue. For a doubling of the shear stiffness of the brain, the strain in the central parts of the brain decreased about 40 percent, and about 25 percent for the brain stem area for an AP rotational motion (Fig. 10). For the same change in properties for a translational AP impulse, a slightly smaller sensitivity is found for the central parts of the brain, while the strain in the brain stem area shows a slightly larger sensitivity than for a rotational impulse.

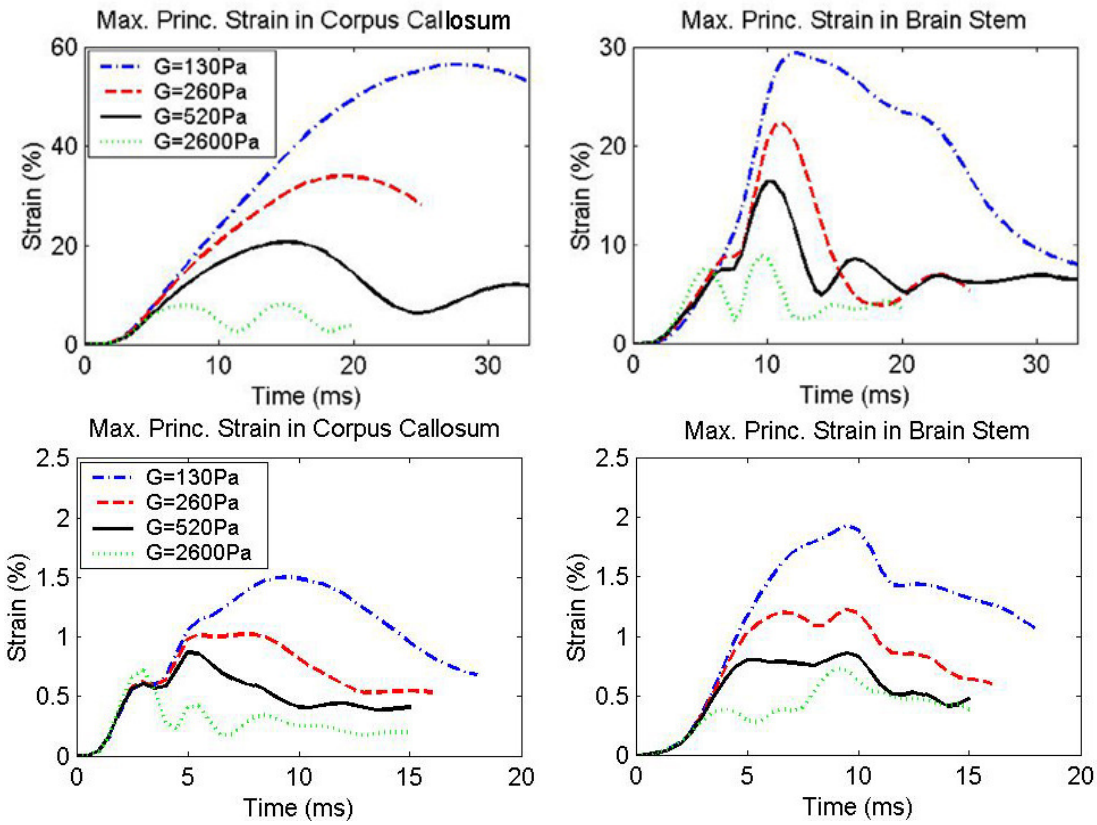

Fig. 10 – Sensitivity of the strain levels to the choice of shear stiffness of the brain. For an AP rotational impulse (upper), and for an AP translational impulse (lower).

DISCUSSION

DIRECTION

The present results verify the hypothesis that a variation in load direction alters the outcome of an impact to the human head. Based on this FE model, new global head injury criteria can be evaluated for all the degrees of freedom of the head. Hence, the injury criteria are valid for a larger span of impact conditions. Injury criteria are today based on a few load directions, but in real life and as indicated by this study, the worst cases for different intracranial components vary depending on the load direction.

The findings of larger stresses and strains in the corpus callosum for the lateral angular acceleration impulse as well as the lateral translational impulse support the conclusions drawn by Gennarelli *et al.* (1982, 1987) that loads in the lateral direction is more likely to cause DAI compared to impulses in the sagittal plane. The largest strains, on the other hand, occurred in the surface of the cortex area. However, large stresses and strains in the surface of the cortex area are related to cortical contusions, and such injuries are usually less critical than the devastating DAI associated with shear strains and effective stresses in the corpus callosum and brain stem areas (Melvin *et al.*, 1993). Strich (1956) 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 shear strain in the white matter adjacent to the cortex 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 (Ivarsson *et al.*, 2000).

Gennarelli *et al.* (1972) subjected 25 squirrel monkeys to controlled sagittal plane head motions, and found brain lesions in both translated and rotated groups but with greater frequency and severity after rotation. This is consistent with the results presented herein, as well as the hypothesis presented by Holbourn (1943). Regarding the translational impulses, larger strains occurred in the spinal cord and brainstem area for the axial impulses (IS and SI) compared to the sagittal AP and PA impulses. For the SI, and the IS translational impulses, the upper part of the spinal cord, and thus the lower part of the brain stem is likely to endure large inertia forces when accelerated in the axial direction. This stretching of the brain stem has previously been discussed in Hodgson and Thomas (1979), who suggested that the mechanism of brainstem

injury, regardless of head motion, is due to shear caused by stretching of the cervical cord. Axial accelerations are usually caused by accidents due to fall and clinical observations shows that this may lead to DAI in the brainstem as well as tearing injuries to the posterior fossa tentorium (Dirnhofer *et al.*, 1979). The findings of high strain in the central parts of the brain and lower strains in the brain stem for the axial rotational impulse supports the findings of Gennarelli *et al.* (1987) that horizontal impulses produce almost exclusively DAI in the central parts of the brain.

EVALUATION OF GLOBAL KINEMATIC-BASED INJURY MEASURES

When it comes to relative motion and strains in the bridging veins, the HIP criterion should give a better prediction of the risk of SDH than HIC. This is evident since the HIP takes into account the load direction and the rotational components of the acceleration. However, the only factors that differentiate between directions in the original HIP are the variation in the mass moment of inertia. Newman *et al.* (2000) therefore proposed a scaling of the impact power for different directions, depending on the tolerance level for the actual direction. The HIP criterion predicts the same levels for the translational impulses as for a corresponding angular impulse, where the highest levels of stresses/strains in the corpus callosum and bridging veins (Kleiven, 2003) are to be found. This gives an indication that weight factors should be introduced to the components of the HIP criterion in order to predict the consequences of impacts where the angular acceleration components are not negligible and a prediction of SDH is desired.

When comparing the different directions, it can be seen that different strains in the brain appear when changing the direction from positive to negative. Thus, the original version of HIP does not distinguish between opposite load direction. In our opinion, three additional components could be added to the original HIP in order to fully take into account the differences in response between opposite directions.

Another problem with HIC and HIP is that they do not seem to capture the level of intracranial response for different impulses. A zero HIC value is predicted for a pure rotational impulse while higher levels of stresses and strains are found compared to a corresponding translational impulse in the same direction. This underlines findings by previous investigators (Gennarelli *et al.*, 1972). One possible explanation could be the as yet unexplored synergic effects of combined loadings. This is included naturally by the product of inertia terms for the angular components in the impact power formulation when using anatomical coordinates. Since the anatomical directions do not coincide with the principal directions of inertia, the product of inertia, I_{xz}, is non-zero. This would add two terms ($I_{xz}\alpha_z \int \alpha_x dt$ and $I_{xz}\alpha_x \int \alpha_z dt$) in equations 6 and 7. In the case of the human head, the power terms containing the products of inertia I_{xy}, I_{xz} and I_{yz} are insignificant compared to the moments of inertia I_{xx}, I_{yy} and I_{zz} (Becker, 1972 and Walker *et al.,* 1973). Nevertheless, these terms in the PI could be estimated using the FE model. In this way, separate scaling factors could be derived to account for synergism of combined directions. In the same manner, supplementary components for the translational terms could also be added to improve the injury prediction.

An almost constant level of strain was found for a constant change in angular velocity, while both the HIP and the peak angular acceleration gave an increasing strain level for an increase in the impulse duration for the AP rotational motion (Fig. 8). This corresponds to Holbourn's hypothesis (Holbourn, 1943) that the strain (and the injury) is proportional to the change in angular velocity for rotational impulses of short durations. For the corresponding translational motion, on the other hand, an almost constant level of strain was found for a constant HIC and HIP, while for a constant change in velocity a decreasing strain level for an increase in the impulse duration occurred (Fig. 9). This supports the results presented by Newman *et al.* (2000), where a good correlation was found between concussion and both the HIC and the HIP for predominantly translational impact data. Since most of the previously proposed angular thresholds are based on non-centroidal rotation in primate experiments followed by analytical scaling techniques, the applicability of thresholds for humans might be discussed. Also, studies on volunteer boxers (Pincemaille *et al.*, 1989) suggest that the human tolerance is largely underestimated using primate experiments and simplistic scaling rules.

BRAIN SHEAR STIFFNESS DEPENDENCY

Biological materials do not follow the constitutive relations for common engineering materials. A biological material is often anisotropic, inhomogeneous, nonlinear and viscoelastic. In addition, there is a great variability between different individuals. The assumption of linear elasticity or viscoelasticity is a great limitation, especially in CNS tissue modeling, due to its typical nonlinear behavior and also because it is often enduring large deformations during impacts and accelerations of the head. Thus, a hyperelastic and

viscoelastic constitutive law was used in the present study. A limitation of this study is the relatively coarse boundary surfaces, differentiating the white and gray matter, and the ventricles. Continuous boundaries have been achieved in 2D (Miller *et al.*, 1998), but never in 3D due to the geometrical complexity of these surfaces. On the other hand, when differentiating between gray and white matter, it becomes somewhat confusing to decide properties for respective tissue. Arbogast *et al.* (1997) reported that gray matter from the thalamus area had about 30 % lower instantaneous modulus than white matter from the corona radiata region, determined by small strain and oscillating shear. The same researchers (Prange *et al.*, 2000) reported that gray matter is slightly (about 4 %) stiffer than white matter when utilizing large strains in simple shear. This illustrates some of the difficulties in determining the properties for brain tissue, as well as the implementation in a numerical model.

The strain in the brain is very sensitive to the choice of stiffness for the brain tissue. When using the parameters reported by Mendis *et al.* (1995), significantly smaller strain was found than when using the average values proposed by Donnelly (1998) and the compliant properties presented by Prange *et al.* (2000). To the authors' knowledge, brain properties roughly half the stiffness of the average published values (around those of Prange *et al.*, 2000) have never been successfully implemented in a simulation before. However, by using such compliant properties a significant increase in the strain in the brain appeared. The characteristics of the response changed as well, producing a larger "delay" in the local brain tissue motion compared to the stiffer and average properties. Recent research suggests that brain tissue is substantially less stiff in extension than in compression and this should also be included in a constitutive model (Miller and Chinzei, 2002).

COMPARISON WITH TISSUE THRESHOLDS

The bulk modulus of brain tissue (McElhaney, 1976) is roughly 10^5 times larger than the shear modulus. Thus, the brain tissue can be assumed to deform in shear. Therefore, distortional strain is 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 (Bain and Meaney, 2000; Bain *et al.*, 1997; Galbraith *et al.*, 1993; Thibault, 1990, Gennarelli *et al.*, 1989), as well as for mechanical injury to the blood-brain barrier (Shreiber *et al.*, 1997). Other local tissue injury measures have also been proposed and evaluated, such as von Mises stresses (Anderson *et al.*, 1999; Miller *et al.*, 1998; Shreiber *et al.*, 1997), product of strain and strain rate (Goldstein *et al.*, 1997; Viano and Lövsund, 1999), strain energy (Shreiber *et al.*, 1997), and the accumulative volume of brain tissue enduring a specific level of strain, the CDSM (Bandak and Eppinger, 1994; DiMasi *et al.*, 1995). For instance, Miller *et al.* (1998) showed in a 2D FE-study that the maximal von Mises stress predicts comparable patterns of axonal and macroscopic hemorrhagic cortical contusions in the miniature pig. Since so many local injury measures are proposed, a thorough evaluation of the correlation between brain injury in humans and some specified local threshold should be performed. When this correlation is achieved, a further evaluation of the required global kinematic-based head injury measure for a specific direction until a certain tissue level is achieved could also be estimated using the FE model.

CONCLUSIONS

Regarding the influence of inertial forces to all the degrees of freedom of the human head, this study shows:

1. HIC is unable to predict consequences of a pure rotational impulse while HIP and peak change in velocity needs individual scaling coefficients for the different terms to account for a difference in load direction.
2. For a purely rotational impulse, the peak change in angular velocity shows the best correlation with the level of principal strain found in the FE model.
3. For a purely translational impulse, the HIC and the HIP show the best correlation with the level of principal strain found in the FE model.

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