THE IMPORTANCE OF ROTATIONAL KINEMATICS IN PEDESTRIAN HEAD TO WINDSHIELD IMPACTS.

Justyna Mordaka[1], Svein Kleiven[2], Margriet van Schijndel-de Nooij[1], Ronald de Lange[1], Luis J.Guerra Casanova[3], Emma L Carter[4], and Hans von Holst[2].

[1]TNO Automotive, P.O. Box 756, 5700 AT Helmond, The Netherlands
[2]KTH, Royal Institute of Technology, School for Technology and Health, Stockholm, Sweden.
[3]Unidad de Biomecánica, Instituto Universitario de Investigación del Automóvil (INSIA), Spain
[4]University of Birmingham, UK.

ABSTRACT

The objective of the present study was to analyze the effect of angular kinematics on head injury in pedestrian head-to-windshield impacts. Three cases of pedestrian head impacts were simulated with FE head and windshield models. The initial impact conditions were obtained from pedestrian accident reconstructions carried out using multi-body pedestrian and car models. The results from the FE head model were compared with injuries reported in the database. Maximum principal strain was chosen as the injury indicator. After successful head injury predictions, the initial velocities were varied and as a result different peak angular velocities and accelerations were simulated. The results showed that increased peak change in angular velocity caused higher maximal principal strain in the brain and in consequence higher probability of Diffuse Axonal Injury (DAI), and Acute Subdural Haematoma (ASDH). A dramatic, three-fold increase in the strain levels in the brain was found when doubling the impact velocity. This paper presents work performed within the framework of a European Commission 6th framework project (APROSYS).

Key words: Brains, Finite Element Method (FEM), Head Injury, Pedestrians.

DURING THE PAST DECADES significant reductions in pedestrian fatalities have been achieved in Europe and the United States. According to the International Road Traffic and Accident Database (IRTAD, 2003), the pedestrian fatalities in 24 European countries decreased from 18410 per year to 7936 per year in the years 1980 to 2001, a reduction of 57%. This tendency is presumably the result of efforts to improve traffic planning and automotive design measures such as changes in car-front shape, improved car-pedestrian stiffness compatibility and anti-lock braking systems (Yang, 1997; Otte, 1999; EEVC, 2002; Berg *et al.*, 2003). Still, injurious car-pedestrian collisions rank high in most countries (Yang, 1997; Berg *et al.*, 2000; EEVC, 2002) and cause head injuries which belong to most frequent and fatal injuries (Yang, 1997; Harruff *et al.*, 1998; Otte, 1999; Mizuno and Kajzer, 2000; EEVC 2002; Mizuno, 2003). Head injuries are almost exclusively caused by the car front end (windscreen, A-pillar, top surface of bonnet/wing) and/or the ground (Yang, 1997; Otte, 1999; Mizuno and Kajzer, 2000; EEVC 2002; Berg *et al.*, 2003; Mizuno, 2003). Therefore, one part of the Euro-NCAP car pedestrian test is the child and adult head impact test and the injury metric is the Head Injury Criterion (HIC), which should be below 1000. 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 (Gennarelli, 1983; Bellora *et al.*, 2001; Kleiven, 2003) while 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. Zhang *et al.* (2004) concluded that both linear and angular accelerations are significant causes of mild traumatic brain injuries. In consequence, the validity of HIC is intensively debated and there is reason to believe that the safety development could be made more efficient by taking into account the effect of rotational kinematics into current safety procedures. The use of more delicate tools, such as biomechanically representative finite element (FE) models of the human head together with local tissue strain thresholds seems to be the best way to evaluate the importance of the rotation. For example, it was found that HIC predicted 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 (Kleiven, 2006a). However, a global criterion will never cover all the various injury mechanisms characterized by local tissue deformation. The aim of the study was evaluate the influence of head rotational kinematics on injury and injury criteria during a typical pedestrian head-windshield impact.

METHODS

HUMAN HEAD FE MODEL

The head model used in this study was developed at the Royal Institute of Technology in Stockholm (Kleiven, 2002). 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). A simplified neck, including an extension of the brain stem into the spinal cord, the dura and pia mater, and the vertebrae was also modeled.

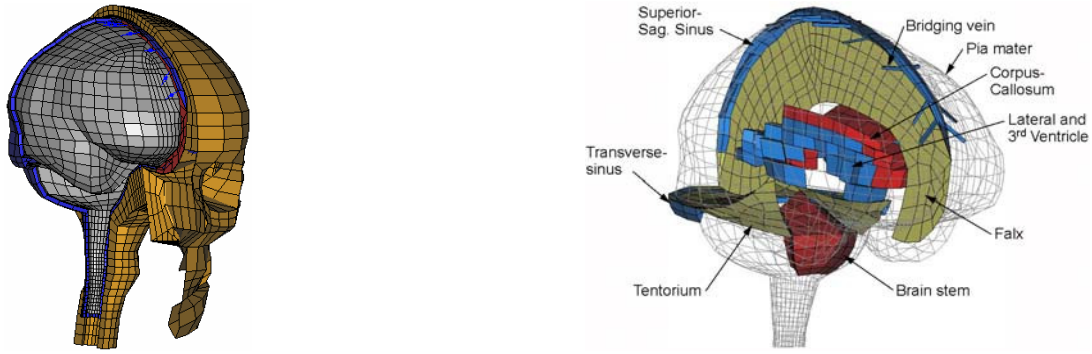


Figure 1 - Finite element model of the human head.

The head model has been validated against several relative motion experiments (Kleiven and Hardy 2002b), intra-cerebral acceleration experiments (Kleiven, 2006a), skull fracture experiments (Kleiven 2006b), and intra-cranial pressure experiments (Kleiven and von Holst, 2002). 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 was used together with a linear viscoelastic model. The constitutive parameters were fitted using discrete spectrum approximation as described by Puso and Weiss (1998). The non-linear elasticity described by Miller and Chinzei (2002), as well as the high frequency relaxation moduli determined by Nicolle *et al.* (2005) was included (Table 1). To obtain the material constants, the iterative least-square algorithm of Levenberg-Marquardt (MathWorks, 1994) was used. Using the relationship $G=\frac{1}{2}\sum \alpha_i \cdot \mu_i$ for the Ogden parameters gives an effective long-term shear modulus of around 1 kPa. These parameters give a quasi-static stiffness for the brain tissue that is around the average published experimental values (Donnelly, 1998). The relaxation moduli, G_i, and Ogden constants, μ_i, were assumed to be 60 % higher than those for the gray matter in the cortex (Arbogast and Margulies, 1997); The stiffness parameters μ_i and G_i were scaled, while the decay constants, β_i, and the Ogden parameters, α_i, were not altered.

Table 1 – Constitutive constants used for the brain tissue.

	Hyperelastic and-viscoelastic constants
μ_1 (kPa)	12.75
μ_2 (kPa)	1600.0
μ_3 (kPa)	-20.00
α_1	1.00
α_2	-0.022
α_3	-1.21
G_1 (MPa)	0.32
G_2 (kPa)	78
G_3 (kPa)	6.2
G_4 (kPa)	8.0
G_5 (kPa)	0.10
G_6 (kPa)	3.0
β_1 (s^{-1})	10^6
β_2 (s^{-1})	10^5
⋮	⋮
β_6 (s^{-1})	10^1

The stress in the cranial bone, maximum principal strain in the brain tissue, change in rotational velocity of the skull, the HIC, the rotational acceleration of the skull, and translational acceleration of the skull for the different windshield impacts 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 (Robbins, 1969,

McElhaney *et al.*, 1970, Wood, 1971) and 30 MPa for the spongeous bone (Robbins, 1969, Melvin *et al.*, 1970) 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 2.

Table 2 - Properties 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.

WINDSHIELD MODEL

The glass model that has been developed by TNO, was chosen to model the car windshield. The glass model consists of three layers: glass-polymer interlayer-glass. The two materials are modeled using different constitutive models. The polymer is modeled using a linear visco-elastic material model. The glass is modeled using a continuum damage model. This model is based on an orthotropic crack model including direction-dependent failure stresses sampled from the failure distribution. This failure probability is Weibull distributed and obtained from experiments on flat glass plates. The windshield model was validated with the experiments of adult head form impacts against a car windshield, which were performed within APROSYS (2006). The computational results were compared to the experiments in terms of the glass crack distribution and head form acceleration and kinematics. The detailed description of the glass model and its validation is explained in Dijk (2007).

SIMULATION SET-UP

Table 3 - Initial conditions (APROSYS, 2005).

Data base reference	Age, gender	Components of head linear velocity (m/s)			Components of head angular velocity (rad/s)			Injury description	Impacted structure
		Vx	Vy	Vz	Ωx	Ωy	Ωz		
IP002	55 yrs, Male, 50% percentile	0.09	2.35	-5.15	8.39	29.0	14.52	Superficial scalp injury AIS=1106021(left side). Stuporous without loss of consciousness level - associated closed head injury AIS=1106022	Car windshield, lower part
IP006	21yrs, Male, 50% percentile	6.8	-0.57	-5.9	6.15	22.1	-2.37	Slight closed cranial-encephalic trauma- AIS 1150999. Haematoma in scalp in occipital area- AIS 1104021 (back side)	Car windshield
BP002	19 years, Male, close to 50% percentile	7.87	0.59	-8.39	9.65	35.5	-9.21	Diffuse axonal injury AIS-146285, Sub-arachn. haemorrhage AIS 1406843, Skull fracture Not sure if head injuries results from windshield or ground impact	Car windshield and secondary ground

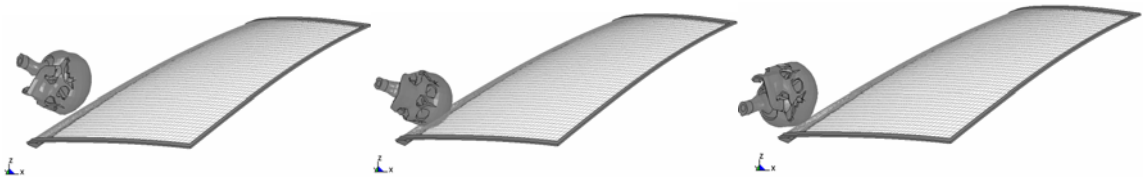
Three documented cases of adult pedestrians impacting car windshields were chosen from an APROSYS database (APROSYS, 2005) and simulated using the FE head and the FE glass model. The

initial conditions and the injury descriptions for the selected cases can be seen in Table 3 and Figure 2. The head orientation and initial linear- and angular velocities are derived from a computational accident reconstruction (APROSYS, 2007) using multibody vehicle and pedestrian models validated for pedestrian impact (van Hoof *et al.*, 2003). The head impact position on the windshield was derived from the documentation and confirmed by the reconstruction. The windshield orientation was based on the pictures from an accident analysis. The injuries description was taken from the medical database. The computational results were checked against existing head injury criteria and compared to the documented injuries. In next step, the initial angular and translational velocities were varied and in consequence different peak angular velocities and accelerations were determined. Several injury mechanisms and related criteria were discussed and their relation with angular kinematics analysed.

RESULTS

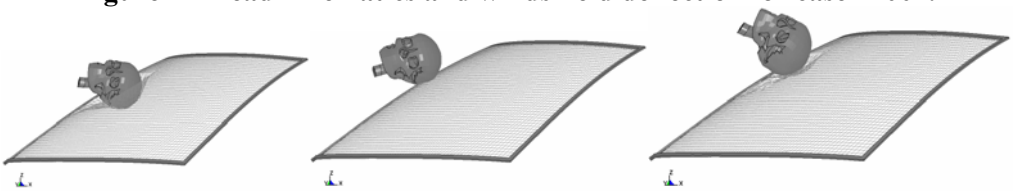
INURY PREDICTION

The kinematics of the three simulations is shown in Figure 2, Figure 3 and Figure 4. Apart from different initial velocities also the initial positions and the head orientations are different for each case which results in different glass deflections and different linear and angular accelerations (Figure 6, Figure 7, and Figure 8 show the kinematics in the global coordinate system). The biggest windshield deflection was observed for case BP002.



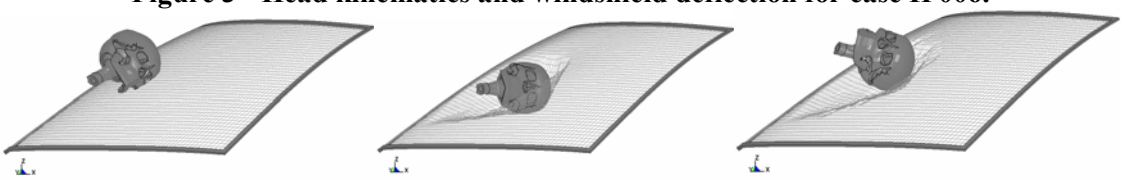
t=0 *t=20 ms* *t=50 ms*

Figure 2 - Head kinematics and windshield deflection for case IP002.



t=0 *t=20 ms* *t=50 ms*

Figure 3 - Head kinematics and windshield deflection for case IP006.



t=0 *t=20 ms* *t=50 ms*

Figure 4 - Head kinematics and windshield deflection for case BP002.

The correlation between damage of windshield from an accident and the model prediction is shown for case IP006 in Figure 5. The cracks in the model spread towards the middle of windshield while in reality they escalate towards the bonnet.

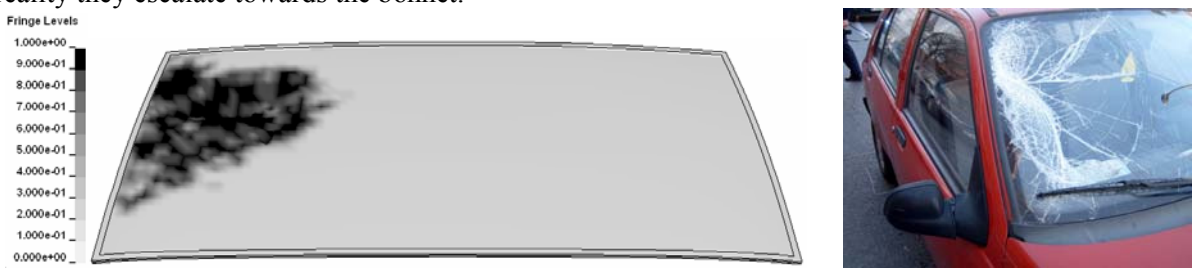


Figure 5 - Visualization of failed elements during the simulation of case IP006 (left) and picture of damaged windshield from the real accident. The failed elements are seen in dark grey.

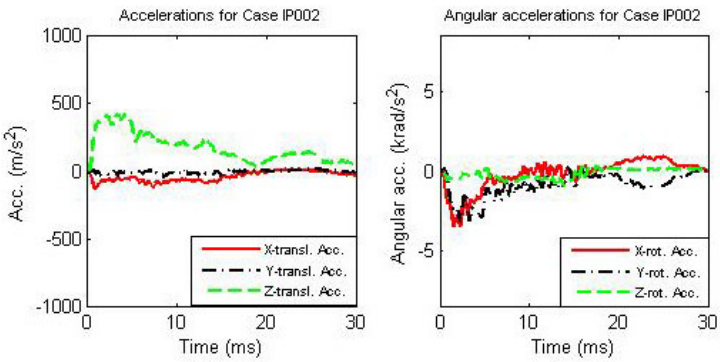


Figure 6 - Head accelerations for the IP002 case.

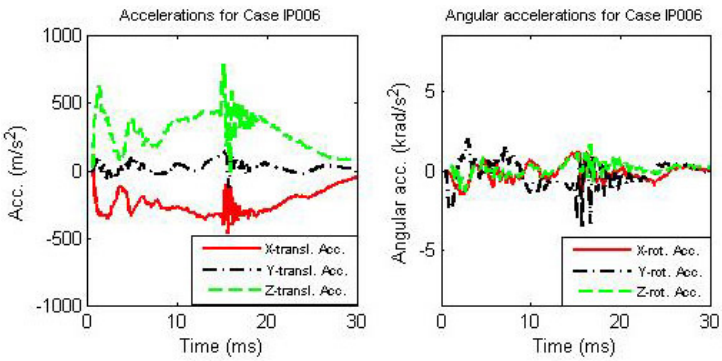


Figure 7 - Head accelerations for the IP006 case.

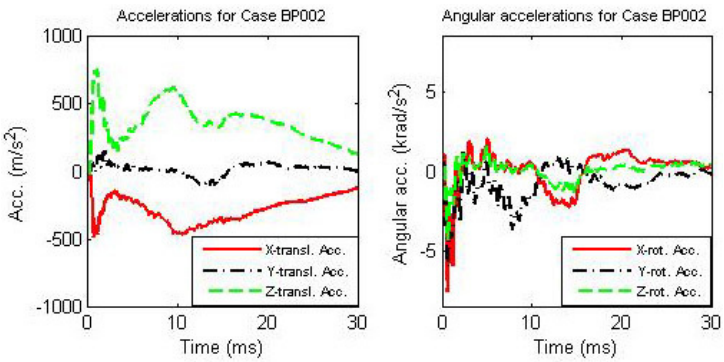


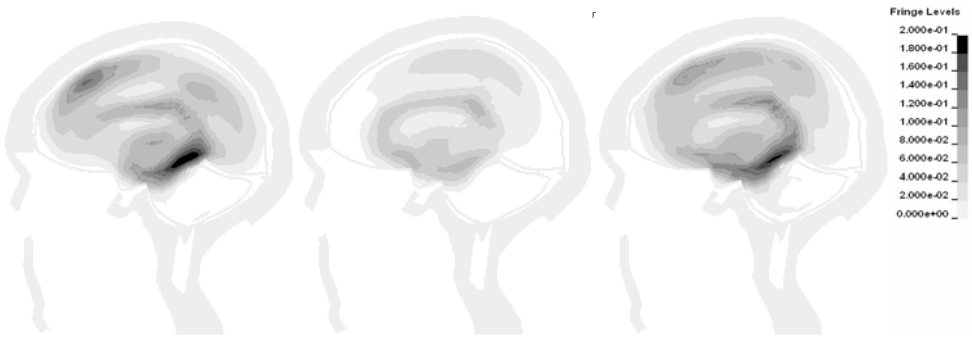
Figure 8 - Head accelerations for the BP002 case.

A summary of the results from the reconstruction of the three different pedestrian windshield impacts using the biomechanical head models is shown in Table 4. Overall, case IP006 and BP002 showed higher values of HIC, peak linear and angular acceleration while higher values in maximal principal strain were observed for IP002 and BP002.

Table 4 - Summary of results from the three different pedestrian windshield impacts.

	IP002	IP006	BP002
Peak acceleration (m/s^2)	428	1109	881
HIC	61	337	475
Peak angular acceleration (krad/s^2)	2,7	3,6	3,0
Peak change in angular vel. (rad/s)	10,6	11,5	22,6
Peak windshield deflection (mm)	29	92	134
Max princ strain in brain	22,9	10,2	21,4
Max princ strain in Corp. Call.	15,0	7,6	12,4
Max princ strain in White M.	17,6	9,9	18,8
Max princ strain in Gray M.	22,9	10,2	21,4
Max princ strain in Br.St.	15,7	9,0	17,4
Max princ strain in Thal./Mid.Br.	15,9	7,5	13,6
von M. stress in outer compact bone (MPa)	5,9	29,8	10,8
von M. stress in inner compact bone (MPa)	2,0	12,1	3,8
von M. stress in por. Bone (MPa)	0,8	10,5	14,4

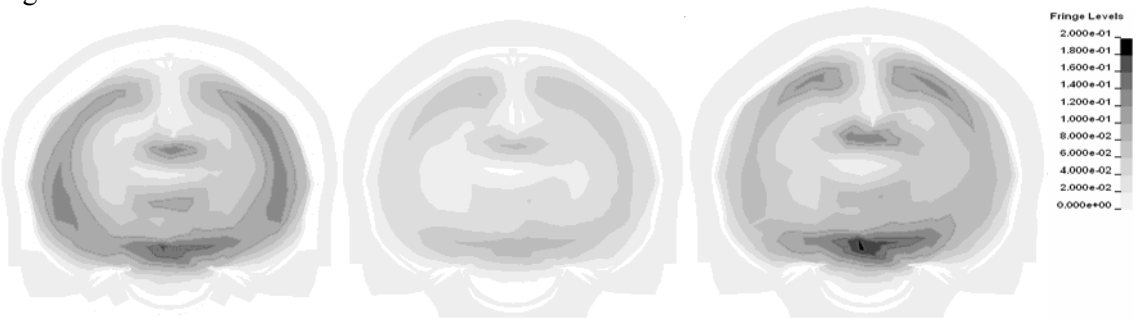
Images showing a parasagittal view of the straining of the brain when enduring the three different pedestrian windshield impacts can be seen in Figure 9. High levels of strain is found in the gray matter close to the brainstem for the IP002 and BP002 cases which resulted in more severe head injuries while the less severe IP006 case also shows lower strain levels in the brain.



IP002 *IP006* *BP002*

Figure 9 - Strain distribution in a parasagittal cross-section (around maximum) for the three pedestrian windshield impacts.

Figure 10 shows the strain distribution in a frontal view of a mid-coronal cross-section for the three pedestrian windshield impacts. High levels of strain is found in the corpus callosum area, as well as in the superior parts of the cortex and in the gray matter close to the brain stem for the lethal BP002 case involving DAI and ASDH. For the IP002 case which involved stuporous without loss of consciousness, high strains can be noted around and close to the brain stem. Correspondingly, for the case IP006 which led to slight closed cranial-encephalic trauma relatively low strain levels are seen throughout the brain.

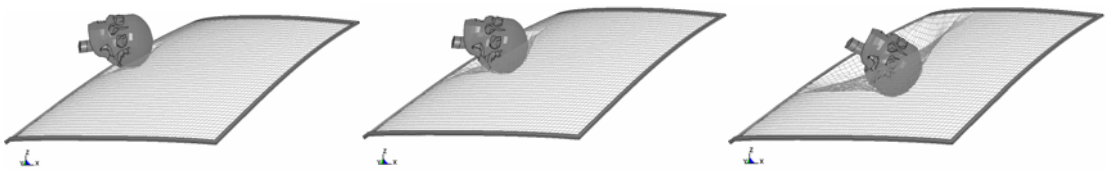


IP002 *IP006* *BP002*

Figure 10 - Strain distribution in a frontal view of a coronal cross-section (around maximum) for the three pedestrian windshield impacts.

PARAMETRIC STUDY

The increase of initial velocity resulted in higher peak change in angular velocity, higher HIC value and lower peak angular and linear acceleration and caused bigger glass deflection (Figure 12, Table 5).



$v=0.5 \times V_{IP006}$ $v=1.0 \times V_{IP006}$ $v=2.0 \times V_{IP006}$

Figure 11 - The maximum deflection of case IP006 for a variation in initial velocity.

It can be seen that when increasing the initial velocity by 100% for the least severe case (IP006), the strain levels in the brain of the human head model increase around than three times (Table 5).

Correspondingly, when scaling the initial velocities to one half, the strain levels are reduced to around half the original value.

Table 5 - Results from the parametric study on the least severe case (IP006) for three different initial velocities.

	IP006 1/2 x vel.	IP006	IP006 2 x vel.
Peak acceleration (m/s^2)	497	1109	1201
HIC	53	337	1724
Peak angular acceleration (krad/s^2)	1,9	3,6	4,7
Peak change in angular vel. (rad/s)	4,3	11,5	25,1
Peak windshield deflection (mm)	38	92	186
Max princ strain in brain	4,6	10,2	31,8
Max princ strain in Corp. Call.	3,3	7,6	20,2
Max princ strain in White M.	4,6	9,9	27,2
Max princ strain in Gray M.	4,5	10,2	31,8
Max princ strain in Br.St.	3,1	9,0	25,1
Max princ strain in Thal./Mid.Br.	3,2	7,5	17,2
von M. stress in outer compact bone (MPa)	7,9	29,8	20,3
von M. stress in inner compact bone (MPa)	1,2	12,1	28,9
von M. stress in por. Bone (MPa)	1,0	10,5	8,6

A different pattern is seen for the stresses in the skull, where the highest value is found for the original, un-scaled velocity. There is a change in kinematics (Fig. 11) where the thinner, left parietal bone endure the highest stresses when using the original velocity, while the thicker skull bone close to the vertex endure the highest stress when doubling the velocity. When using half the original velocity, the occipital part of the left parietal bone endures the highest stress.

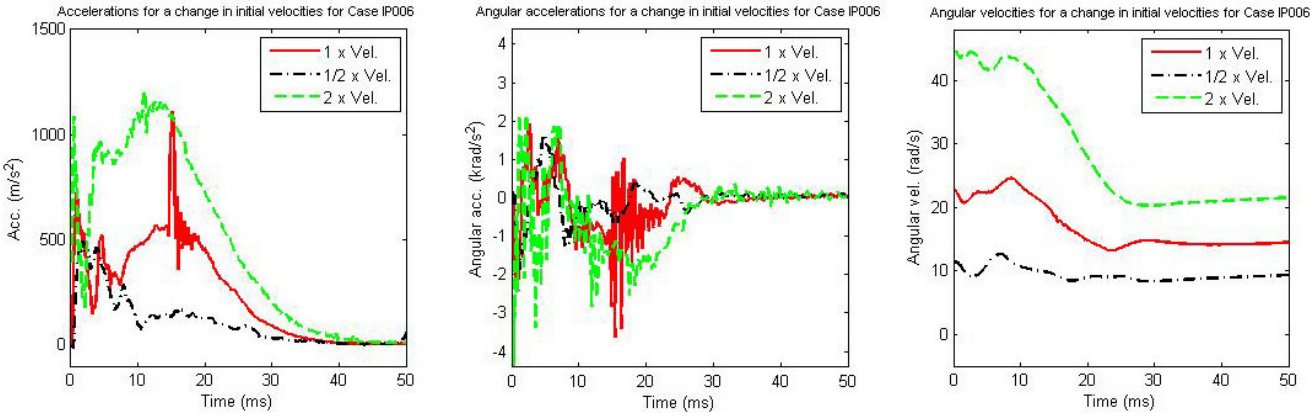


Figure 12 - Head accelerations and angular velocity for a variation in initial angular and translational velocity.

DISCUSSION

Table 6 - Real injury and prediction.

	IP002		IP006		BP002*	
	Real injury	Model prediction	Real injury	Model prediction	Real injury	Model prediction
Skull fracture	no	no	no	no	yes	no
SDH	no	no	no	no	yes, severe	no
DAI	yes, moderate	yes	yes, minor-mod.	no	yes, fatal	yes
Contusion	yes	yes	no	no	yes	yes
Concussion	yes	yes	yes	yes	yes	yes

*The severe head injuries for the BP002 case is, according to the accident analysis, most likely due to the un-simulated ground impact.

The medical description of the real injuries and AIS codes were analysed by an experienced neurologist and the head injury severities were attempted to be expressed in terms of concussion,

contusion, DAI, ASDH and skull fracture. For each of the injury types a criterion was assigned and then compared against the head model results (Table 6).

The bulk modulus of brain tissue (McElhaney *et al.*, 1976) 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 (Bain and Meaney, 2000, Morrison III *et al.*, 2003, Gailbraith *et al.*, 1993, Thibault *et al.*, 1990) 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 stress (Shreiber *et al.*, 1997, Anderson, *et al.*, 1999, Miller *et al.*, 1998), the product of strain and strain rate (Goldstein *et al.*, 1997, Viano and Lövsund, 1999, King et al, 2003), the strain energy (Shreiber *et al.*, 1997), and the accumulative volume of brain tissue enduring a specific level of strain, the Cumulative Strain Damage Measure (CSDM), (Bandak *et al.*, 1994, DiMasi *et al.*, 1995). 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 (Kleiven, 2006c, Kleiven, 2007). This strain is very sensitive to the choice of stiffness for the brain tissue (Kleiven and Hardy, 2002) and more work is needed to fully describe the non-linear and viscoelastic response of living brain tissue. However, the chosen properties for the constitutive model is close to the effective shear modulus of around 10 kPa at 80 Hz found for brain tissue *in vivo* by McCracken *et al*. (2005) using magnetic resonance elastography (MRE). The constitutive model is sophisticated including both non-linear elasticity, which distinguished between tension and compression, and high frequency viscoelasticty using six relaxation moduli and decay constants. However, when sustaining inertial forces, the brain is mainly distorted in shear and tension so differentiating between tension and compression might not be essential. Also, high frequency viscoelasticity might not be needed since the impacts do have relatively constant time durations of around 20-30 ms for most cases which could be well covered with one relaxation modulus and time constant.

It is suggested that maximal principal strain higher than 18-21% leads to DAI (Bain and Meaney, 2000; Morrison *et.al*, 2003) while the vascular rupture is expected at strain levels above 30-60% according to Lövenhielm (1974) and Monson (2003). For case IP002, the maximal principal strain reached values about 20% which corresponds to slight risk of DAI (close to the brain stem area) and excludes the chance of SDH. From the database it is known that the pedestrian had endured moderate severity (AIS-2) closed head injury which means that there was no open wound or other connection to the external area through the skull bone. The medical report suggests that the patient suffered concussion, moderate DAI and contusions. The model predicted concussion since studies performed with giant squid axons (Thibault, 1993) suggested a maximal principal strain of around 10% to cause reversible injury to the axons which could be used as an approximate threshold for concussion. Moreover, the strain level in the cortex area indicated a high risk of contusion based on Shreiber *et al.* (1997) who derived a threshold of 19% in principal strain in the cortex for a 50% risk of cerebral contusions.

The medical description indicates that the pedestrian IP006 suffered cranial-encephalic trauma with an AIS-9. Cranial-encephalic trauma means that the victim has received a trauma to the head without specifying skull or brain tissue injury while the AIS-9 simply means that no AIS injury severity was provided. However, the injury description said: Slight closed head injury and occipital haematoma in the scalp, which suggest that the correct injury severity would be AIS-1/2. It could be interpreted that this victim suffered a severe concussion on the border to moderate DAI, and correspondingly low maximal principal strains (around 10%) were found in the brain model which indicate concussion.

The case BP002 was the most severe in terms of brain injury since both DAI and ASDH were reported. The FE model indicates high strains (above 20%) in the brain which corresponds to a risk of axonal damage. However, since lower strain levels than 30% were found in the cortex, the model did not predict any risk of vessel rupture and ASDH. McElhaney *et al.* (1970), Melvin *et al.*(1970), Robbins *et al.(*1969) and Wood (1970) have reported cranial bone stress thresholds. According to the mentioned references, compact cranial bone breaks in tension at 48-128 MPa, while the cancellous bone breaks in compression at 32-74 MPa. Skull fracture was reported for BP002. However, the FE model, based on these criteria did not predict skull facture. On the other hand, it should be noted that

the skull fracture for BP002 is on the opposite side of the contact point of the head with the windshield and therefore the severe head injuries are most likely caused by the ground impact. This is in line with the accident database in which a violent impact with ground was mentioned. This fact indicates that also the DAI and SDH might have been a result of the ground impact which was not simulated in the present study. The highest strain in the brain was found for case IP002 although this case had the lowest HIC, peak change in angular velocity and angular acceleration. This emphasizes the need of accounting for the impact direction in global injury criteria. It is worth noticing that the IP002 and BP002 cases are predominantly enduring loading in the coronal and axial planes, while the least severe case, IP006, is mainly loaded in the sagittal plane. These findings 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. Since the HIC values in the three simulated cases were below 1000, the predicted risk of injury according to that criterion is low which does not correspond to the injury level describe in the database.

The parametric study using a variation of the initial velocities emphasize the importance of the impact speed for the injury outcome. A dramatic, three-fold increase in the strain levels in the brain was found when doubling the impact velocity. This variation in strain in the brain for a difference in impact speed is varying similarly to the peak change in angular velocity. This corresponds to Holbourn's hypothesis (1943) that the strain (and the injury) is proportional to the change in angular velocity for rotational impulses of short durations. Margulies and Thibault (1992) 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 (Margulies and Thibault, 1992). For the highest velocity in the parametric study of the IP006 impact in the present study, an angular acceleration of 4.7 krad/s^2 and a peak change in angular velocity of 25.1 rad/s was found together with a maximum strain in most parts of the brain exceeding 20 percent. On the other hand, the HIC is not insignificant for this impact. It has previously been found that HIC and the head impact power (HIP) (proposed by Newman et al., 2000) predicted 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 (Kleiven, 2006a). Recently, it was also found that a simple linear combination of rotational velocity and HIC showed a high correlation with the maximum principal strain in the brain (Kleiven, 2007). Therefore, a combination of the peak change in angular velocity and HIC could probably predict the difference between impacts of various severities. In this parametric study, impact at only one location of the head is studied and therefore the results will most likely also differ depending on what impact location that is chosen.

The current results must be treated with special caution due to several estimations and assumptions that have been made. The head initial orientation and direction and values of velocities were derived from accident reconstructions which aimed to reproduce the global pedestrian kinematics and final position and not exact head kinematics. Using these parameters might introduce some discrepancy into the simulations. Also the exact head impact points on the windshields were not known and they were estimated based on photos of the damaged windshields and computational reconstructions. Moreover, the geometry of the FE windshield models are taken from different car models and only its angle relatively to ground was adjusted to represent the profile of the specific car that was used in the accident. However, the maximum deflections for the IP002 and IP006 cases corresponds quite well to the experimental values found in drop tests using free motion head forms (APROSYS, 2006). Another assumption, which could influence the head model response, was using a 0.4 friction coefficient for head-glass interaction which is usually used for head form-car bonnet impact simulation. Finally, although the database includes the head injury descriptions, it is not sufficient and accurate enough to determine with confidence the injury type, severity and location. It would be ideal to have the medical images of the brain injuries or report from autopsy.

CONCLUSIONS

The KTH FE head model was used together with a detailed FE glass model to simulate head-windshield impacts experienced in real pedestrian accidents. These simulations demonstrated the

capability of the head model of distinguishing between pedestrian head to windshield accidents of different severities by means of maximum principal strain in the brain and skull fracture by von Mises stress. The parametric study showed that head rotational kinematics had great influence on injury severity, particularly on DAI and should therefore not be neglected. Moreover, the severity of injury was found to be related not to angular acceleration but to peak change in angular velocity. Further parametric studies are needed to fully understand the relation between head impact location/orientation, and a relevant injury criterion.

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