Influence of Impact Velocity and Angle in a Detailed Reconstruction of a Bicycle Accident

Madelen Fahlstedt, Katrien Baeck, Peter Halldin, Jos Vander Sloten, Jan Goffin, Bart Depreitere, Svein Kleiven

Abstract Bicycle accidents have become the most common cause of serious injury in the traffic during the last couple of years in Sweden. The objective of this study was to investigate the effect of the input variables, initial velocity and head orientation, of a bicycle accident reconstruction on the strain levels in the brain using a detailed FE head model. The accident involved a non-helmeted 68 year old male who sustained a linear skull fracture, contusions, acute subdural hematoma, and small bleeding at the swelling (subarachnoid blood). The orientation of the head just before impact was determined from the swelling appearing in the computer tomography (CT) scans. The head model used in this study was developed at the Royal Institute of Technology in Stockholm. The stress in the cranial bone, first principal strain in the brain tissue and acceleration were determined. The model was able to predict a strain pattern that correlated well with the medical images from the victim. The variation study showed that the tangential velocity had a large effect on the strain levels in the studied case. The strain pattern indicated larger areas of high strain with increased tangential velocity especially at the more superior sections.

Keywords Accident reconstruction, bicycle, finite element method, head injuries

I. INTRODUCTION

Bicycle accidents have become the most common cause of serious injuries in the traffic during the last couple of years in Sweden [1]. In 2009 one third of the road users admitted to hospital were cyclists. In Belgium and Sweden, around 10% of all traffic fatalities are cyclists [2]-[3]. A high proportion of the serious injuries in bicycle accidents are to the head [4]. Depreitere et al. [5] found that more than 70% of the 86 bicycle accident cases studied, involved skull fracture, contusion and brain swelling whereas [6] found soft tissue to be the most common head injury followed by concussion.

Computer models, such as MADYMO and Finite Element (FE) models, can be used to better understand the mechanisms behind an injury and for example be used in reconstruction of real-world accidents. The knowledge gained from these reconstructions can in turn give important inputs for development of safety equipment. Accident reconstructions can be very useful but there are difficulties associated with an accident reconstruction due to the limited information about the accident circumstances. A small variation in one parameter, e.g. impact velocity or impact location, have shown in several pedestrian accident studies to give a large variation in the results from the simulations [7].

Some accident reconstruction studies of bicycle accidents have been done but not much has been published on the effect in output when the input is varied. A fall from a bicycle can appear in many different ways and the variation of the kinematics, head orientation and velocity can be large. A multibody program such as MADYMO

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has the advantage of being computationally efficient but FE models can better describe the injuries at tissue levels. A detailed FE model can also be used to study the mechanic effect on tissue level. In previous studies of bicycle accidents, MADYMO has been the dominating working tool. Therefore, the objective of this study was to investigate the effect of different input variables on the strain levels in reconstruction of a bicycle accident with a detailed FE model. The variation study included initial velocity and head orientation.

II. METHODS

The starting point for the variation study was an accident reconstruction done in MADYMO, presented in [8]. The accident involved a non-helmeted 68 year old male (1.82 m and 70 kg). He was riding downhill at approximately 20 km/h and got stuck with the front wheel of his bicycle in a small gap between two bricks in the road. He was then thrown over the steering wheel. The victim sustained a linear skull fracture (left temporal), contusion (hemorrhagic right temporal), acute subdural hematoma (ASDH) (diffuse and scattered and small strip next to the contusion), and small bleeding (subarachnoid blood) at the swelling site.

The head orientation at impact was determined from the swelling appearing in the computer tomography (CT) scans. The midsection of the swelling is shown in Fig. 1. In Fig. 2 the initial orientation of the head model is shown. The variation study of the head orientation included a change of ±10 degrees around global z-axis, +10 and +20 degrees around y-axis, and -10 and -20 degrees around x-axis.

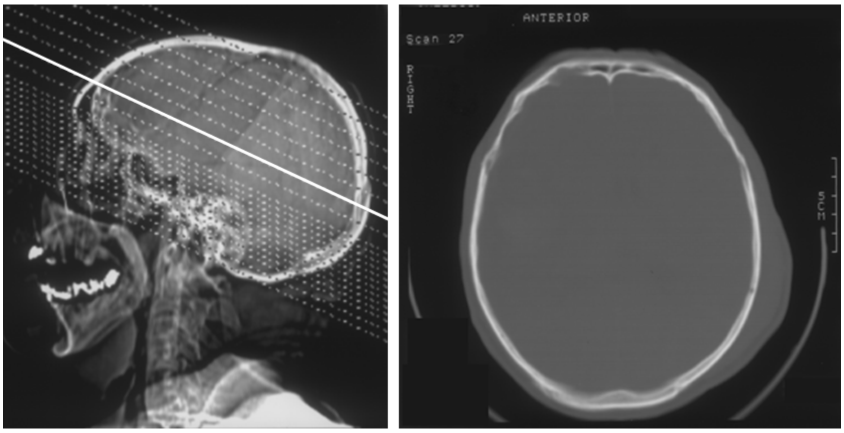

Fig. 1. The medical image indicating the midsection of the swelling. The solid lines in the left figure indicating the scan's level.

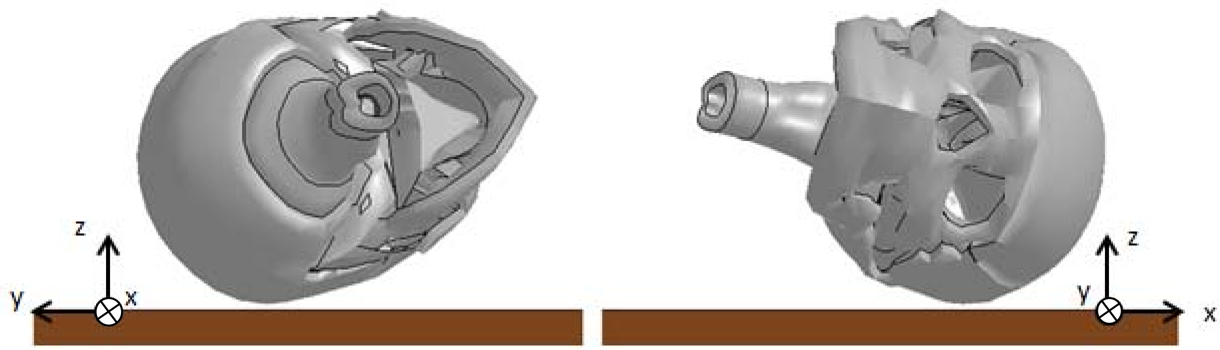

Fig. 2. The initial head orientation.

The accident reconstruction generated from MADYMO gave the resultant linear velocity of the head just before impact of 5.3 m/s (v_x= 3.4 m/s, v_y = 0.75 m/s and v_z =-4.0 m/s) and resultant rotational velocity of 4.7 rad/s. Verschueren [8] presented ten accident reconstructions of falls from bicycles. The vertical initial velocity varied between -4.0 and -5.4 m/s (standard deviation (SD)= 0.43) whereas the tangential velocities varied between 1.06 and 8.58 m/s (SD = 2.59). In the variation study the velocity components of x and z were varied ±1 SD. The ground was modeled as elastic material twice as stiff as the skull bone. The contact between the ground and the head was modeled with a penalty based contact. The friction coefficient was set to 0.4.

The FE head model used in this study was developed at the Royal Institute of Technology in Stockholm [9]. 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. A simplified neck, including an extension of the brain stem into the spinal cord, the dura and pia mater were also modeled. More detailed information about the model and validations can be found in [10]-[11].

To cope with the large elastic deformation of the brain, a third order Ogden hyperelastic constitutive model was used together with a linear viscoelastic model. The constitutive constants used for the brain tissue are presented in Table I, and more information about the constants can be found in [11]. To evaluate the effect on the brain tissue the Green-Lagrange 1st principal strain was used.

TABLE I
HYPERELASTIC AND VISCOELASTIC CONSTANTS FOR THE BRAIN TISSUE

μ_1 (Pa)	53.8
μ_2 (Pa)	-120.4
α_1	10.1
α_2	-12.79
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

μ_i – Ogden constants; G_i – relaxation moduli; β_i – decay constant

To account for the possible loss of load bearing capacity at high contact loading, the stresses in the skull were limited to 80 MPa for the compact bone [12] and 30 MPa for the spongeous bone [13]-[14] through the use of an isotropic constitutive model with plasticity. A summary of the properties for the tissues of the human head used in this study is presented in Table II. The von Mises stress in the cranial bone was analyzed. The linear acceleration, angular acceleration and angular velocity were also evaluated. The acceleration and velocity curves were filtered with an SAE 1000 filter.

TABLE II
PROPERTIES USED IN THE NUMERICAL STUDY

	Young's Modulus [MPa]	*Density [kg/m^3]*	*Poisson's ratio*	*Yield Stress [MPa]*
Outer compact bone	15 000	2.00	0.22	80
Inner compact bone	15 000	2.00	0.22	80
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	-	
Sinuses	K = 2.1 GPa	1.00	-	
Dura mater	31.5	1.13	0.45	
Falx/Tentorium	31.5	1.13	0.45	
Scalp	Hyper-Viscoelastic	1.13	0.42	

K - Bulk modulus
More information about the constants can be found in [9]

The skull bone has been further validated for this study with experimental data from [15]. They did experimental tests on Post Mortem Human Subjects (PMHS) where a 12 kg rigid plate was impacting the temporal-parietal region at 4.3 m/s. The difference in force-displacement between the experiment and the computer model is shown in Fig. 3.

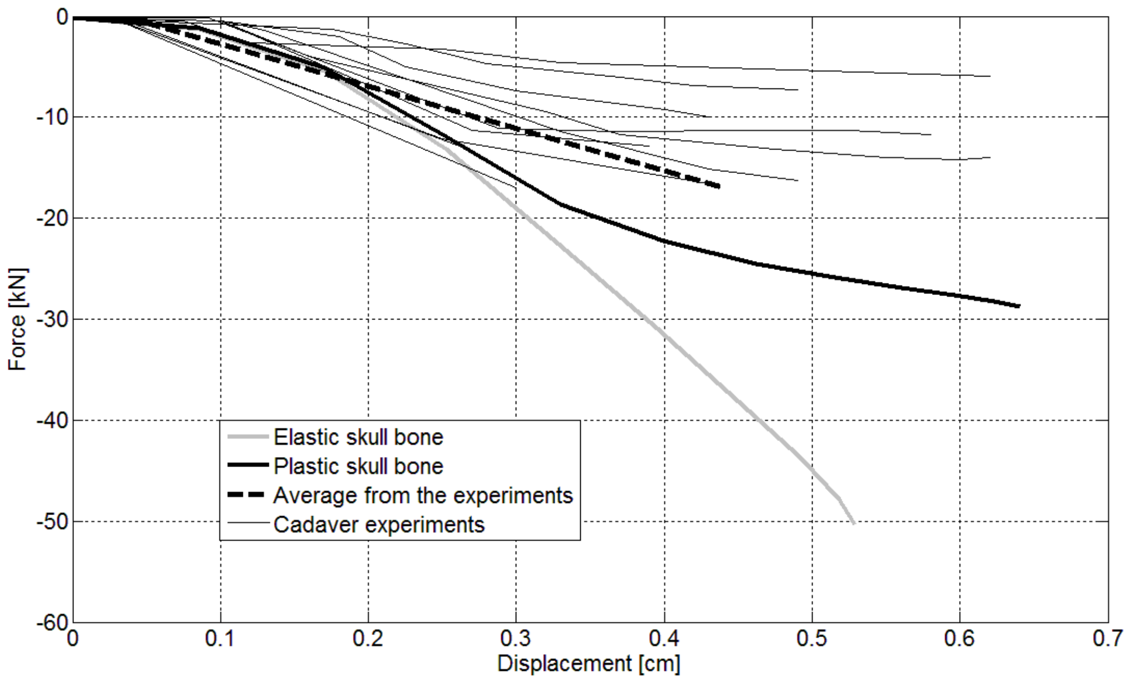

Fig. 3. Force versus deflection, comparing the simulation and the experiment.

III. RESULTS

The strain distribution in the brain tissue from the initial input variables is presented in Fig. 4 together with the corresponding medical images. The section plots were taken when the maximum strain of all sections occurred. The maximal 1st principal strain in the sections of the corresponding medical images is presented in Table III.

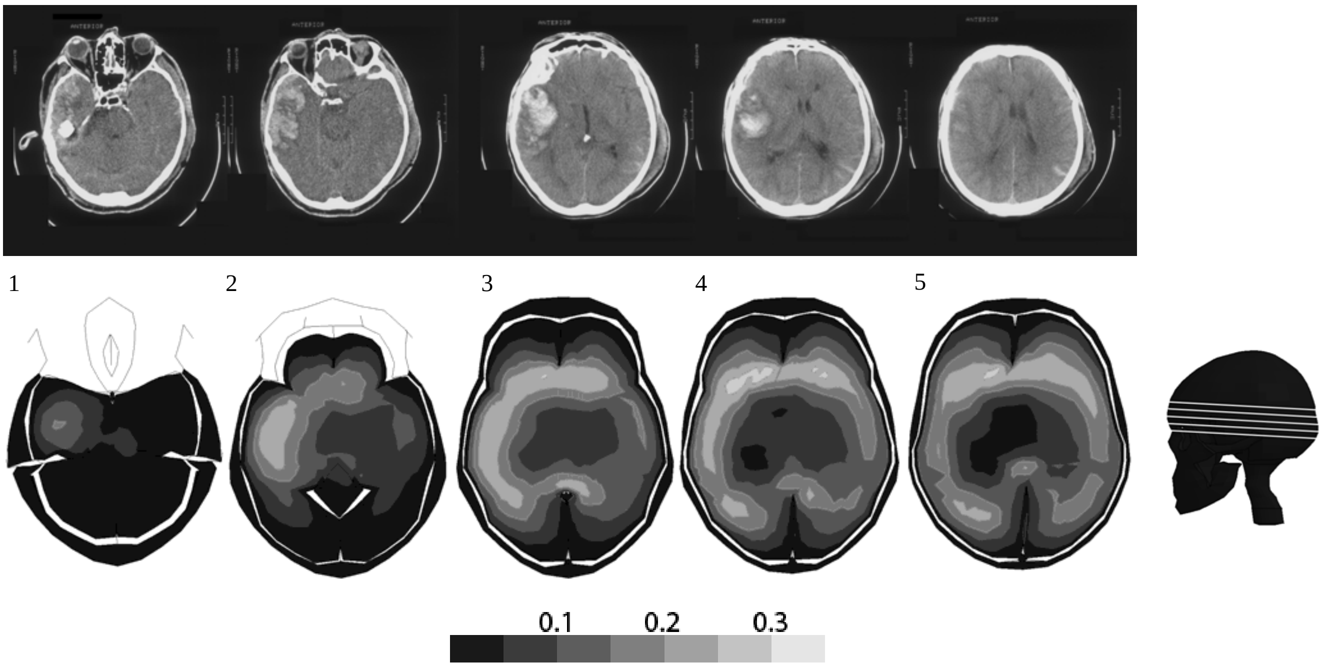

Fig.4. Comparison between the medical images and the strain pattern from the initial simulation. The sections are numbered 1 to 5 from left to right. The sections' levels for the model are shown in second row to the right.

TABLE III
THE MAXIMAL 1st PRINCIPAL STRAIN IN THE SECTIONS FROM THE INITIAL SIMULATION

	Section 1	*Section 2*	*Section 3*	*Section 4*	*Section 5*
Initial simulation	0.24	0.28	0.29	0.29	0.27

The von Mises stress level in the compact bone reached plasticity (80 MPa) in all cases. The plastic zone changed with the variation of head orientation and initial velocity. The plastic zone for the simulation with the initial parameters is shown in Fig. 5. The variation of the 1st principal strain for the five different sections when the velocity and orientation were varied is shown in Fig. 6 to 7. The strain pattern for these simulations is shown in the Appendix. The peak linear acceleration, angular acceleration and change in angular velocity for the different simulations is shown in Table IV, and the time history plot is shown in the Appendix.

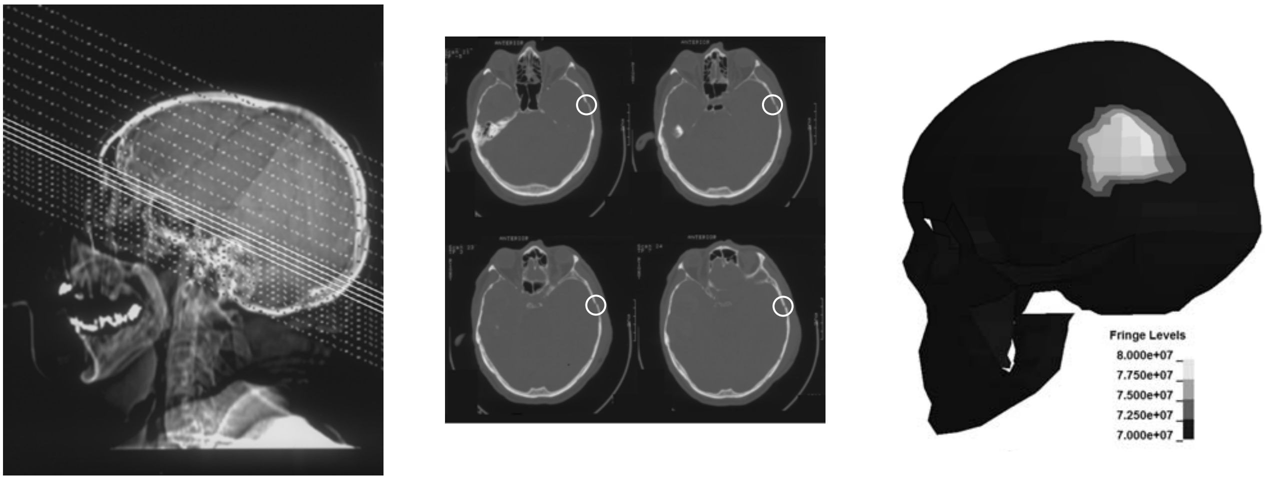

Fig 5. The medical image showing the fracture and the plastic zone from the initial simulation. The solid lines in the left figure indicating the scans' level.

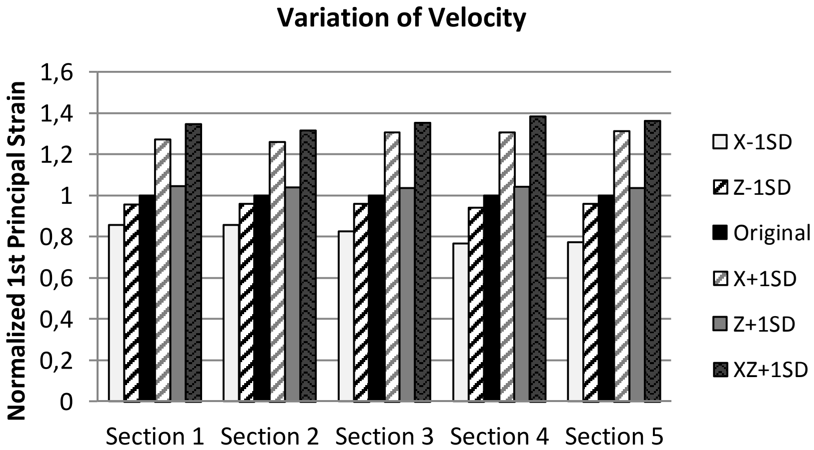

Fig. 6. The normalized max 1st principal strain in the sections variation with different velocity.

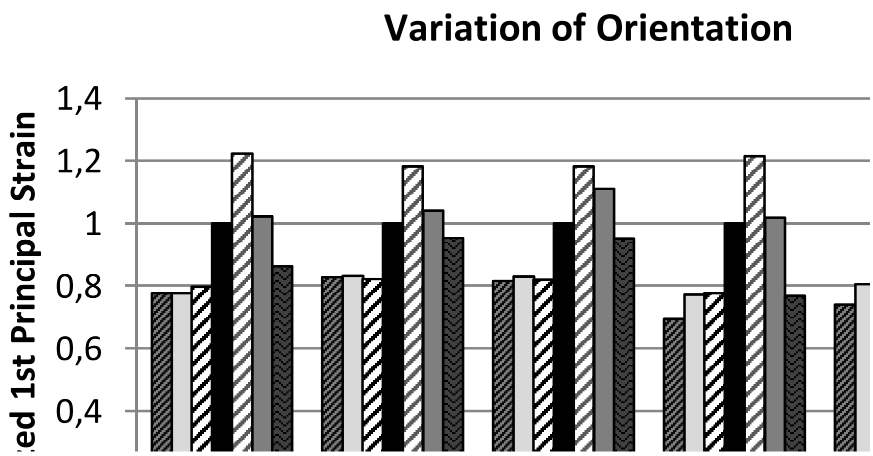

Fig. 7. The normalized max 1st principal strain in the sections with different head orientation at impact.

TABLE IV
PEAK CHANGE IN ANGULAR VELOCITY, PEAK LINEAR ACCELERATION, PEAK ANGULAR ACCELERATION AND MAXIMUM CONTACT FORCE FOR ALL CASES

	Peak change in angular velocity [rad/s]	Peak linear acceleration [m/s^2]	Peak angular acceleration [krad/s^2]	Contact force [kN]
Initial	24.9	4350	22.6	20.8
-1 SD X-velocity	19.1	4380	24.8	21.7
+1 SD X-velocity	38.3	4990	32.8	20.8
-1 SD Z-velocity	24.3	4160	22.7	21.6
+1 SD Z-velocity	26.0	5260	24.7	19.7
+1 SD XZ-velocity	39.2	4960	29.1	21.9
Rotation -10Z	20.0	4603	18.9	20.7
Rotation +10Z	30.9	4490	27.8	20.7
Rotation +10Y	31.0	4540	25.4	20.7
Rotation +20Y	28.4	5080	18.7	23.8
Rotation -10X	23.2	4700	18.3	21.6
Rotation -20X	24.7	4580	19.1	22.0

IV. DISCUSSION

This paper has investigated the strain level in a FE model of the human head during the reconstruction of a bicycle accident. The model was able to predict a strain pattern that correlated well with the medical images from the victim. The strain level in the model was between 0.24-0.29 which is covering the experimentally found level of around 0.2 for a 50% risk of contusions reported earlier [16]. The threshold for ASDH is between 0.3-0.7 for strain failure of the vessels according to [17]-[19]. The strain levels of the tissue can seem to be a little low to predict ADSH but the thresholds can also be affected by the age of the victim, in this case an older man. The largest injury areas in the brain scans are located in the scans 2 and 4. These levels are also the sections with highest strain in the initial simulation.

The original impact location was determined from the swelling in the CT scans since no more detailed information about the impact location was known. That the bleeding was caused by the initial impact is not certain but most likely. The location of the bleeding can give an important input for an accident reconstruction.

The stress level in the skull reached a plasticity stress of 80 MPa according to experimental data from [12]. The highest stress area was located at the impact location posterior from the fracture line. The fracture could have appeared due to bending. The model used in this study has a rather coarse mesh and no failure criterion. To study the fracture in more detail, a model with finer mesh and failure criterion should be used. In this study an isotropic constitutive model with plasticity was used to model the load limiting due to yielding and fracture of the compact and spongeous bone. In the validation against [15] the difference between the plastic and elastic skull bone was seen after 1.5 mm displacement. The plastic skull bone had a more similar stiffness after 1.5 mm displacement then the elastic skull bone compared to the average from the experiments. The displacement of the skull bone was 2-3 mm in the simulations.

The variation study showed that the tangential velocity had a large effect on the strain level. The maximal strain increased from 0.29 to 0.40 when the tangential velocity was 1 SD higher. An increase was seen in all sections. The strain pattern indicated larger areas of high strain with increased tangential velocity especially at the more superior sections. The highest strain was found at the right side, as in the medical images. When the tangential velocity was decreased the highest strain was more found to the left and more inferior. The strain level was also lower, varying from 0.20 to 0.24.

The variation of vertical velocity gave a small difference in maximum strain. The maximum strain varied from 0.23-0.27 for the different sections when decreasing the vertical velocity and 0.25-0.30 for increased vertical velocity.

The velocities in vertical and tangential direction have been varied with the standard deviation from the MADYMO simulations [8]. Therefore the percentage change in velocity is larger for the tangential velocity. The vertical velocity varied between -3.6 and -4.4 m/s corresponding to a falling height of 0.7 to 1.0 m. The vertical velocity can seem a little low. The realistic maximum vertical velocity in bicycle accident would be close to the velocity created in a fall from standing height.

By increasing the head rotation around the global z-axis the maximum strain increased 16% to 22% for the sections. The opposite effect was seen when decreasing the rotation around z-axis, -18% to -24%. The reason for the big influence of the Z-rotation is probably due to the fact that the position of the head center of gravity is changed. The tangential force prevents head rotation motion when the z-axis rotation is decreased due to the position of the head center of gravity. When the rotation is increased the tangential force is restricting the rotation less.

The rotation around y-axis with 10 degrees gave a small increase of maximum strain (-1% to 10%) but 10 degrees further gave a decrease of maximum strain (-6% to -25%). The areas with strain over 0.2 were decreased in both cases compared to the initial simulation but most in 10 degrees rotation. The rotation around x-axis gave a decrease of maximum principal strain. The decrease was rather equal for 10 and 20 degrees in section 1 to 3, -18% to -23%, but a slightly larger decrease was seen for 20 degrees in the superior sections.

The peak in linear acceleration varied between 424g and 536g whereas the maximum contact force was 18.3-23.8 kN. This is high but in all simulations the stresses were high, all cases reached the plasticity of 80 MPa. Strain values have shown to correlate well with change in angular velocity [11]. A change in peak angular velocity followed the change of maximum strain compared to the initial simulation and which was seen in all simulations except when the head was rotated +20 degrees around y-axis. In the same way the peak in angular acceleration followed the strain in all simulations except when the tangential velocity was decreased.

There is a three order of magnitude difference between the brain's shear modulus and its bulk modulus so that for a given impact the brain tends to deform only in shear. This also gives a large sensitivity of the strains to rotational loadings and a small sensitivity to translational kinematics. The strain magnitude in the brain is sensitive to only the rotational kinematics and not the translational motion [11].

In this study the rotation around x- and y-axis was just changed in one direction. This was due to the fact that it was unrealistic to rotate the head in the opposite direction because it would give an almost lateral impact. The shoulder and rest of the upper body is inhibiting a strictly lateral impact. This is a limitation of the study since no account is taken to the effect of the neck, shoulder and the rest of the upper body. Other studies [8],[20]-[21] have investigated the effect of upper body and suggested that the impact duration, impact velocity and impact location determines the importance of the upper body.

In this study the maximum principal strain at the first 20 ms after impact in the five sections and the strain pattern at the time with maximum strain were evaluated. It could be interesting to study the change of strain during the impact. It was also seen that areas superior to the sections presented in this study, close to the impact location, had high strain also but this was not included in this study.

However, even if there are a number of limitations, the simulations show that it is possible to capture the strain pattern that is similar to the medical images. The study has shown that the FE head model can be used for variation studies in order to improve the knowledge about head injuries. This can be used to improve the design of better head protection.

V. CONCLUSIONS

The objective of this study was to evaluate the effect on strain levels in the brain when input parameters, like velocity and orientation, were changed. In this particular bicycle accident case, the tangential velocity showed to have the largest effect on the strain levels.

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VII. APPENDIX

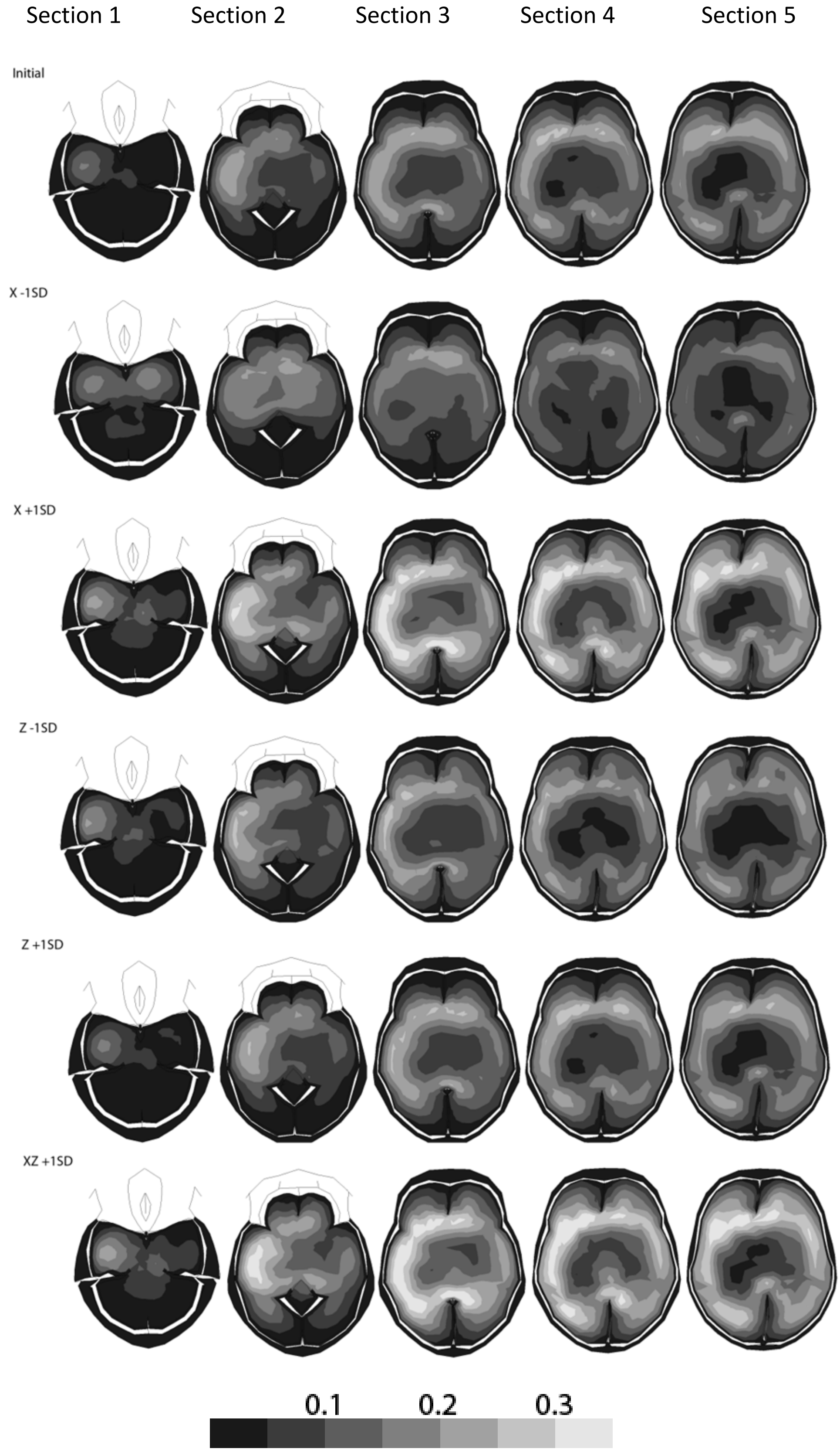

Fig. A.1. Strain pattern when the vertical and tangential velocities were changed.

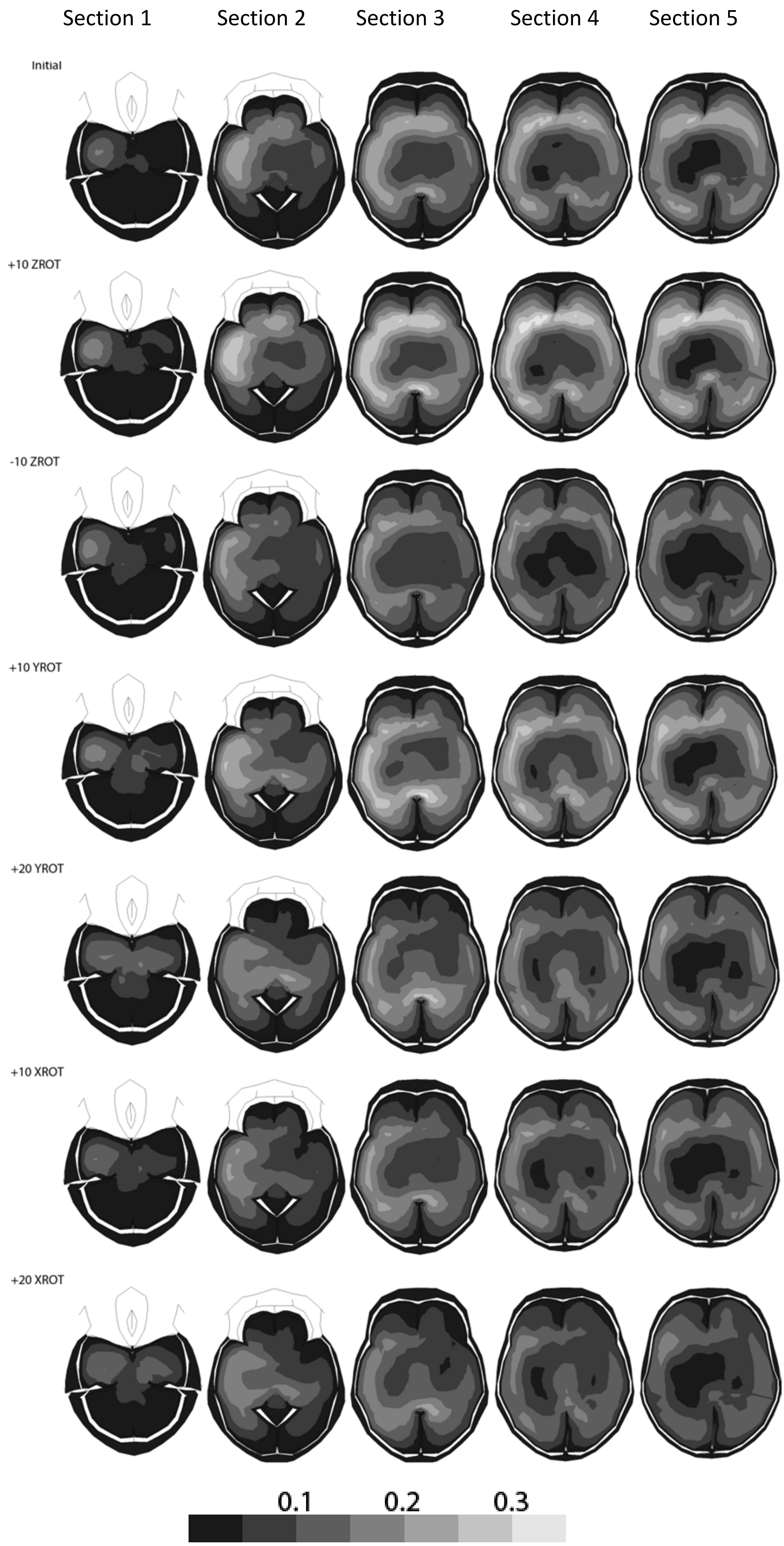

Fig. A.2. Strain pattern when the head orientation was changed.

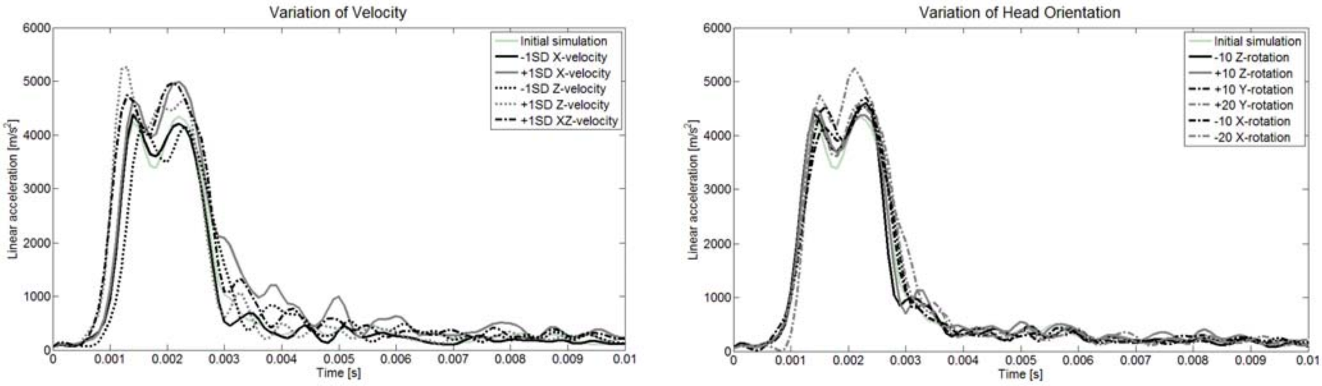

Fig. A.3. Linear acceleration with different velocity (left) and different head orientation (right).

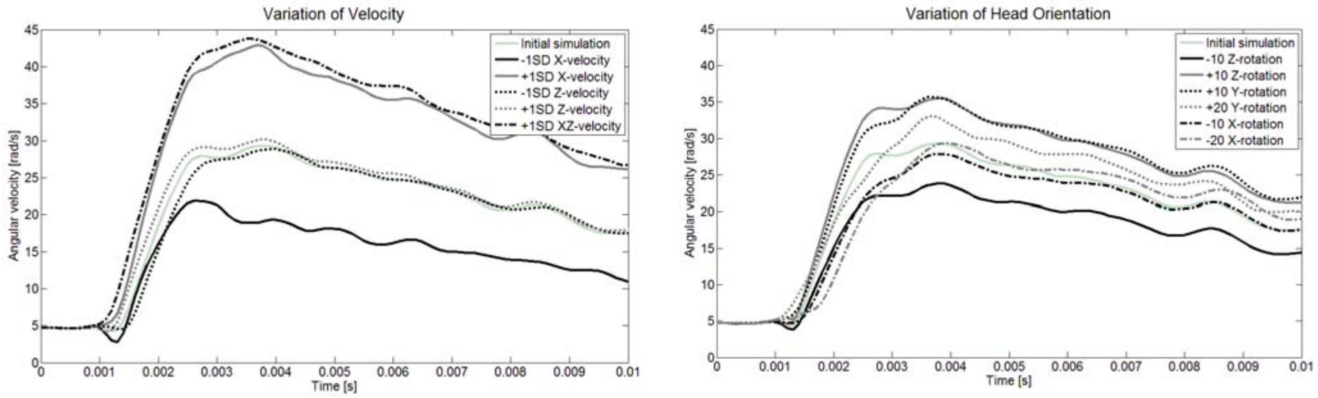

Fig. A.4. Angular velocity with different velocity (left) and different head orientation (right).

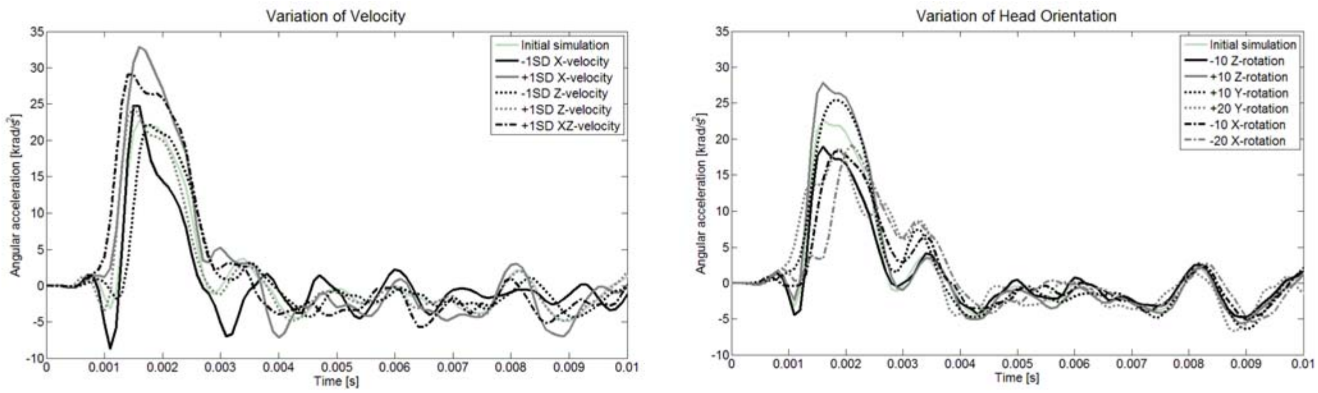

Fig. A.5. Angular acceleration with different velocity (left) and different head orientation (right).

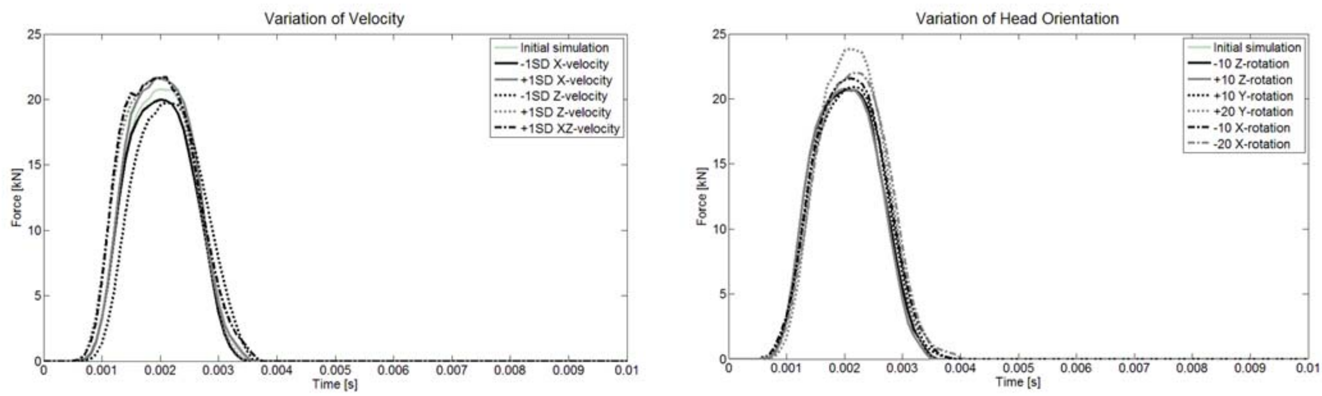

Fig. A.6. Contact force with different velocity (left) and different head orientation (right).