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Using Finite Element simulations to reproduce rotational induced brain injury experiments: Recommendations for the future

Master's Thesis in the Applied Mechanics

JOEL HULTMAN

Department of Applied Mechanics
Division of Traffic Safety, Injury Prevention Group
CHALMERS UNIVERSITY OF TECHNOLOGY
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ABSTRACT

The response of the rat brain during rotational acceleration was investigated using finite element simulations to improve understand of the mechanisms responsible for injuries in the brain. Finite element simulations were carried out to match the animal experiment carried out by *Davidsson et al* (2009). The results from the simulations were compared to injuries inflicted by a rotation of the brain in the animals during the animal experiment. The rat brain was modelled as a viscoelastic, isotropic material; the animal brain model was subjected to a rotational acceleration to mimic the experiments. Different parameters, such as displacement, shear and von mises stress, were investigated and a number of different approaches to model these experiments were investigated. Findings from the simulations show a correlation between shear stress and the location of injuries seen in the animal experiments. The pressure from the simulation was extracted. However, problems were present during the pressure extraction. The solution to this mesh related problem; i.e. changing tetrahedral elements to hexahedral elements, could not be achieved during the given time span of the thesis. Validation methods, i.e. pressure and deformation, were investigated. It was concluded that it will be almost impossible to validate the material stiffness of the brain tissue during the experiment from deformation of the tissue. Since the deformation of the small rat brain will, due to its size, not give a measurable change in displacement for the stiffness tested in the simulations.

Key words: Finite Element, Rat brain, Animal simulation, Viscoelastic, Rotational acceleration, Validation Data.

Contents

Preface

During this thesis a number of simulations have been done in LS-Dyna to investigate the behaviour of a rat brain subjected to rotational acceleration. Studying trauma using animal models can give a deeper understanding of the mechanisms that introduce injuries to the brain. Such understandings will hopefully also increase the understanding of how injuries to the brain are caused in the human brain as well. The thesis work started in the middle of January 2010 and was carried out until the end of May 2010 at the Department of Applied Mechanics, division of Injury Prevention.

Acknowledgements

I would like to thank *Karin Brolin* for her expertise and guidance in finite element method and LS-Dyna, *Kristian Holmqvist* and *Jonas Östh* for their introduction to LS-Dyna and help during the thesis work, *Johan Davidsson* for his guidance during this thesis work. And all the others at SAFER who helped me throughout the work with this thesis. A special thank to *Patrik Wåhllin* for his Proofreading of this thesis.

Göteborg,
Joel Hultman

Notations

Beta APP	beta-amyloid precursor protein is normally transported in the axons but accumulates at the site of injury to the axon and is thereby used to trace damage to the axons
DAI	Diffuse Axonal Injury
FE	Finite Element
LS-Dyna	Finite element software from Livermore Software Technology Corporation
TBI	Traumatic brain injury

1 Introduction

Karolinska Institutet and Chalmers have jointly carried out animal experiments to investigate the injury to the brain inflicted by rotational acceleration. With the development and use of various mathematical models an improved understanding of how these injuries occur will be obtained. Based on this and other studies a new injury criterion can be developed.

A commonly used method in biomechanics is the *finite element method*. An important part when using *finite element models* in injury biomechanics research is validation of the models. Since gathering experimental data is time consuming and restricted to what can be measured it is of interest to investigate different approaches for validation of these finite element models. Such a study can guide researchers in their selection of approach for the validation of the models.

1.1 Background

Each year millions of people are brought to hospitals with a traumatic brain injury. For all the advances in medical science, injuries to the brain are still difficult to treat. Traumatic brain injury is therefore a leading cause of death and disability worldwide *Ninds* (2010).

In about half of the traumatic brain injury cases *diffuse axonal injuries*, DAI, are present. The diffuse axonal injury is categorized as a diffuse brain damage, i.e. an injury that occurs over a wider area inside the brain; specifically for the DAI is the injury to the axons. Severe diffuse axonal injury is one of the leading causes of death in people with traumatic brain injury *Ninds* (2010).

In the experiment done by *Davidsson et al* (2009) the diffuse brain injuries from rotation was investigated using living rats. In this thesis the same experiments were reproduced using numerical simulations.

1.1.1 Evaluation of the model

There are a limited number of ways to evaluate a numerical model. Some of the limitations in this thesis are related to the small size of a rat brain used in the experiment done by *Davidsson et al* (2009), where it is hard to get measurements from the experiment, i.e. a small relative change in a small brain will lead to a even smaller absolute change. Possible measurements for evaluating the numerical model can be:

- Injury presence and injury pattern
- Displacement of the brain tissue and strain in the tissue
- Pressure in the ventricles or the brain tissue

This list only includes what can, generally, be measured and not how to measure these physical quantities. Injuries can relatively easy be observed after an experiment. However, pressure, displacement and strain are not as easy to measure in an experiment of this scale and duration.

1.1.2 Numerical evaluation

Other properties that can be used for evaluation of the numerical model are:

- Hourglass energy
- Eroding elements
- Geometry of the model compared to a real rat brain
- Boundary conditions
- Contacts between different parts of the model

To validate the numerical model, the ratio between the total energy and hourglass energy can be used for evaluation. The model should not have a hourglass energy that is higher than 5% of the total energy. If the ratio between total energy and hourglass energy is higher excessive numerical deformations and instabilities will be present. If this ratio is not kept at a reasonable level the simulation will deform. This deformation will not be due to the applied force but to the numerical error of the hourglass energy.

An option used during the simulation was "*erode element*"; which is a method used, in LS-Dyna, to deletes elements that collapses (i.e. when an element get negative volume) instead of terminating the simulation. Eroding elements can introduce errors to the calculation, because the element is removed from the model after it has been eroded and therefore will decrees the stiffness locally in the model.

To set up a representative numerical model of an experiment the mesh should describe the shapes of the object that is studied. The element size and element shape will affect the stiffness of the model. Therefore a convergence study should be done to eliminate the effects of the element size. This was not done due to limitations in the software used to create the mesh. However a comparison of different element types was done! The layer of the *Cerebrospinal fluid*, CSF, surrounding the brain does not give a good representation of the real CSF since the numerical CSF is four times too thick; this is discussed in the Discussion section.

The boundary conditions for the model of the rat brain should describe the conditions inside the brain during the trauma. The parts inside the brain are tied together with shared nodes. This is a good way to speed up and make the calculations stable, since otherwise a number of contact algorithms must be used.

1.2 Aim

The overall aim of this thesis was to study diffuse brain injury during sagittal plane rotational acceleration by combining experimental and numerical models. And give recommendations regarding validation data to validate the numerical model.

The more specific aims of the study were:

- Adopt an existing animal brain FE-model and reproduce experiments, in which the animal head was exposed to rotational trauma, to improve the understanding of brain trauma.
- Investigate correlation between brain damage and the dynamic pressure, stress and strain in the brain parenchyma as predicted by the FE-model.
- Investigate methods to validate the FE model.

1.3 Limitations in the thesis

On limitation in the current work was that the materials have been modelled as being isotropic while the material in the brain is not non isotropic. The non isotropic aspect of the material will not be addressed during this thesis because of the lack of material data regarding the non isotropic behaviour.

The software available during this thesis was *LS-Dyna*, *HyperWorks* and *Simpleware*. *Simpleware* was used to create the mesh of the experiments. The control over the mesh generation in this software was limited. Therefore a mesh containing only hexahedral elements could not be created during the given time span.

2 Literature

A literature survey was conducted in the beginning of this thesis work to guide the design of the study. The current knowledge about DAI, FE-Models and validation methods related to this thesis was investigated.

Lauret (2006) suggest that strain is the cause for injuries to the brain tissue. She concluded based on a literature review that a rapid stretch of axons can damage the axonal cytoskeleton, and thereby induces an injury to the axon. However, *Zhou et al* (1997) saw a strong correlation between shear strain and DAI. They suggest that it will be possible to give a prediction of DAI from the shear contours.

Kleiven (2002) shows that many brain models are not enough thoroughly validated for the purpose of the models. Out of 9 models that he investigates only 4 are validated and 3 of them are only validated against the pressure. *Bradshaw and Morfey* (2001) concluded that finite element models validated for dynamic internal pressure cannot be assumed to be validated for injury prediction. Since strain or shear is not closely related to pressure.

Shulyakov et al (2009) determined the material properties for an adult rat brain in vitro and ex vitro. The elastic properties were determined with a controlled deformation to the brain by a vinyl screw that was inserted into the brain. The Elastic modulus was determined to be 31 ±2 [kPa] and poissons ratio of 0.49. However, they show that brain tissue exhibits viscous properties.

In *Brands* (2002) the Viscoelastic material properties are validated from measuring the deformation of the tissue with neutral density markers, in the range 0 to 20% shear strain. However outside of this range the material properties are not validated. The material properties used for humans are; bulk modulus *2.5* [GPa], shear modulus range *250-3300* [Pa]. Pressure history data was used to validate this model as well as 3D x-ray markers, to capture the deformation. Oscillatory shear experiments were used to validate the strain, but they also maintain that the deformations seen in a crash (20% and above) cannot be validated.

2.1 Injuries extracted from the experiment

To gather information on the injuries on the brain during rotational acceleration Davidsson et al (2009) exposed a number of animals to a rotational load in the sagittal plane of the head. After trauma these animals were dissected and the brain tissue stained for injury detection. *Beta APP* was used to stain where the concentrations of these elements were carefully mapped.

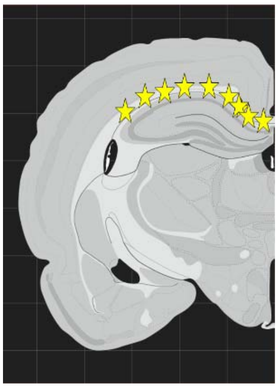

Figure 1) The stars in the figure indicate the axon injuries mapped by Davidsson et al (2009). This is the injury that the model results will be compared to.

Positive Beta-APP axons were seen between the cortex and the corpus callosum, as shown in *Figure 1*. All animals subjected to a peak acceleration of *1.1* Mrad/s^2 showed concentrations of Beta-APP positive axons in the same regions. The levels of Beta-APP positive axons were subjectively determined according to a grading scheme, seen in Davidsson et al (2009).

A higher concentration of the *Beta-APP* can be seen in *Figure 2*, where an injured axon is seen. This will provide a good indication of where the injuries occur and therefore the *Beta-APP* findings will be used for the comparison between the experimental results and those obtained in the simulations.

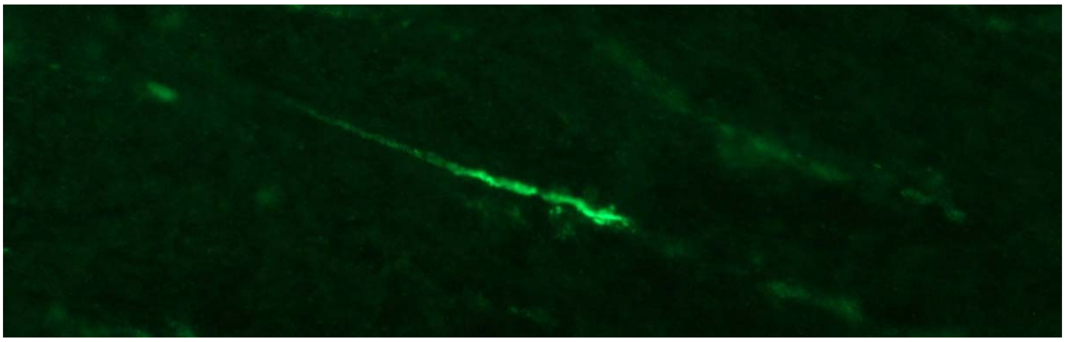

Figure 2) Showing a high concentration of the Beta APP protein along a axon or dendrit. The length of such a concentration is a typically few micrometers. Beta-APP is gathered along an injured axon or dendrit which therefore can show where the injuries to the axons occur.

2.2 The finite model of the rat brain

The model consists of a mesh of the brain, CSF and skull bone originally created by *Gianfranco* (2010) assigned different material properties and boundary conditions. The material data is taken from the literature survey. The boundary conditions were set to represent the experiment done by *Davidsson et al* (2009).

2.2.1 Parts of the rat brain included in the model

The mesh of the skull and rat brain, adopted and used in this study was originally created by *Gianfranco* (2010. In *Gianfranco*'s thesis a mesh of the skull, CSF layer and rat brain was created with a number of images from a MRI-scan. The mesh consists of a skull bone layer and CSF layers, and a number of parts in the rat brain, see *Figure 3*.

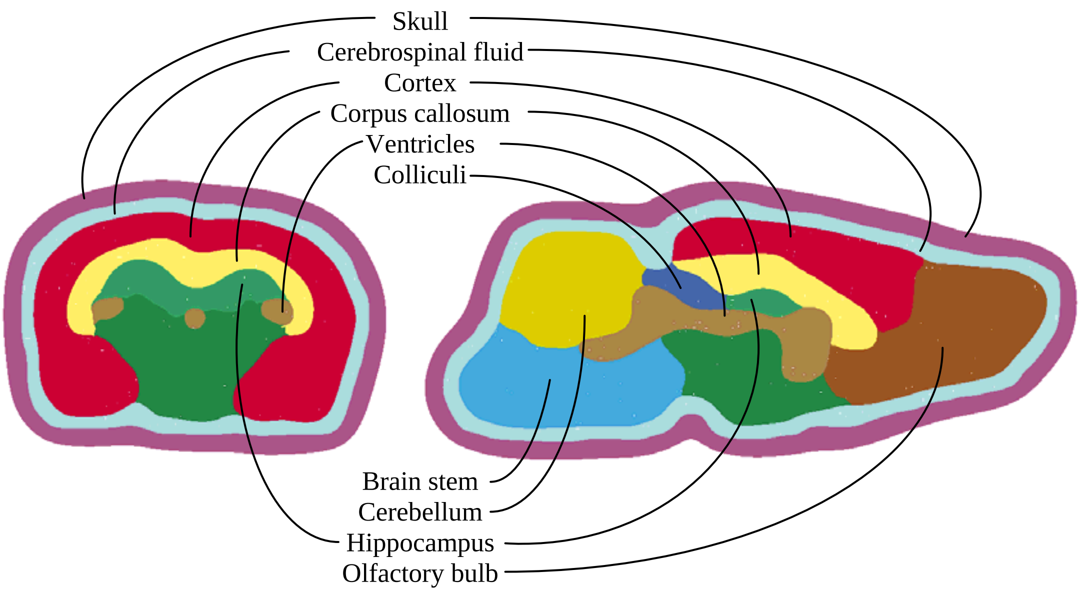

Figure 3) a figure showing different parts of the brain.

2.2.2 Material data and model

Different mathematical material models have been used during simulation of brain material in the past. In the first developed finite element models of the brain, elastic material models were used. Currently the trend is to use more complex material model such as viscoelastic material models (*Zou* 2007)

During the literature survey different material data and material models were investigated. For the final selection of brain and CSF material data and material model, a rat specific data and corresponding model was selected. The material data that selected comes from *Haojie et al* (2006) and can be seen in *Table 1*. The material model selected to describe the brain were a linear viscoelastic model, where in LS-Dyna 971 the corresponding material model was called Linear_viscoelastic_brain.

Table 1) The viscoelastic materials used in the simulation, taken form Haojie et al (2006). These materials all share the same bulk modulus of 2.19 [GPa] and a density of 1040 [kg/m^3] and the long and short term shear modulus can be seen in the table.

Viscoelastic materials			
	Short term [Pa]	**Long term [Pa]**	**Decay constant [ms]**
Cerebral gray matter, Cerebellum, and Brainstem	1720	510	20
White matter	1200	360	20
Ventricles	1000	300	20
Spinal cord	3100	920	20

The material properties assigned to the model can be seen in *Table 2*. The properties can be seen in *Table 1*.

Table 2) The properties assigned to the parts in the model.

Part in the model	Property
Brainstem	Spinal cord
Caudate putamen	Gray mater
Cerebellum	Gray mater
Colliculi	Gray mater
Corpus callosum	White matter
Cortex	Gray mater
Hippocampus	Gray mater
Olfactory bulb	Gray mater
Ventricles	Ventricles

Other parts of the head that can be found in the model are *Cerebrospinal fluid* (CSF) and skull bone. Both of these were modelled using elastic material. This was in line with the recommendations by *Bourdin et al* (2007).

The CSF material was assigned an elastic modulus of 0.1 [MPa], Bulk modulus of 2.19 [MPa], poisons ratio of 0.45, a viscous factor of 0.05, density of 1040 [kg/m^3] and with the option solid as fluid turned on in LS-Dyna 971. This will, according to Bourdin et al (2007), be a good assumption for the CSF. For future information about *solid as a fluid*, see *LS-DYNA 971 theory manual* (2006).

A large range of the elastic module for the skull bone material have been used in simulations, with a range from 0.2 – 20 [GPa] according to *Kleiven* (2002). However, in these simulations the skull bone was model as a stiff elastic material; with an elastic modulus of *15* [GPa], density of *2000* [kg/m^3] and the poisons ratio was *0.22,* values taken from *Kleiven* (2007).

3 Materials and Methods

A finite element model was set up to numerically simulate the experiment carried out by *Davidsson et al (2009)* with the goal to provide a numerical model that could describe the animal experiment.

3.1 Simulations setup

A number of different simulations were carried out to investigate the injury mechanism responsible for the injuries observed in the rat brains following rotational trauma. Simulations were carried out to compare the numerical outputs, such as stress strain and pressure, from the models to the given injuries from the experiments.

Using LS-Dyna 971 as the solver for the numerical simulations the effect of a number of different parameter changes, such as element type, stiffness, boundary conditions, centre of rotation, on the injury predictability were investigated, these parameters was chosen to show what type of data that is needed for the validation of the model. The simulations were then analysed, by comparing different simulations it was concluded how the model was affected by the changes to the model.

Under-integrated elements, i.e. only one integration point in the element instead of in each node, are used during the simulation, therefore the hourglass energy need to be monitored for a valid simulation. There are different ways to handle the hourglass problems related to under-integrated elements:

- Hourglass control (add stiffness and dampening to the element)
- Fully integrated elements (there cannot be any hourglass energy)

Fully integrated elements are resource intense, compared to the under-integrated elements. Since the hourglass effect only occurs when elements are under-integrated the hourglass problem will automatically disappear when fully integrated elements are used. However, new problems will arise, such as volume instability. The simulations could not be completed with fully integrated elements.

To minimize the calculation time and reduce negative volume problem, i.e. collapsing elements, hourglass control schemes is used together with under-integrated elements instead of fully integrated elements. As long as the hourglass energy is less than 5% the model is acceptable, at least with respect to the hourglass energy. When hourglass control is used a numerical stiffness is added to the material which could result in a model that is stiffer than it was originally assigned to be; according to the material data. All of the simulations produce a ratio of hourglass energy that was lower than *5%*.

3.1.1 Boundary conditions

Two boundary conditions were used. Both these prescribe a motion of a node set during the simulation. The prescribed motion is calculated from the rotational acceleration curve seen in *Figure 4*. The prescribed motion of the nodes selected for these boundary conditions are then rotated around the centre of the rotation as seen in *Figure 5*. One node set is put on the skull bone to mimic the rod glued to the skull in the real experiment to introduce the rotational acceleration and the other node set is on the edge of the spine to follow the motion of the skull.

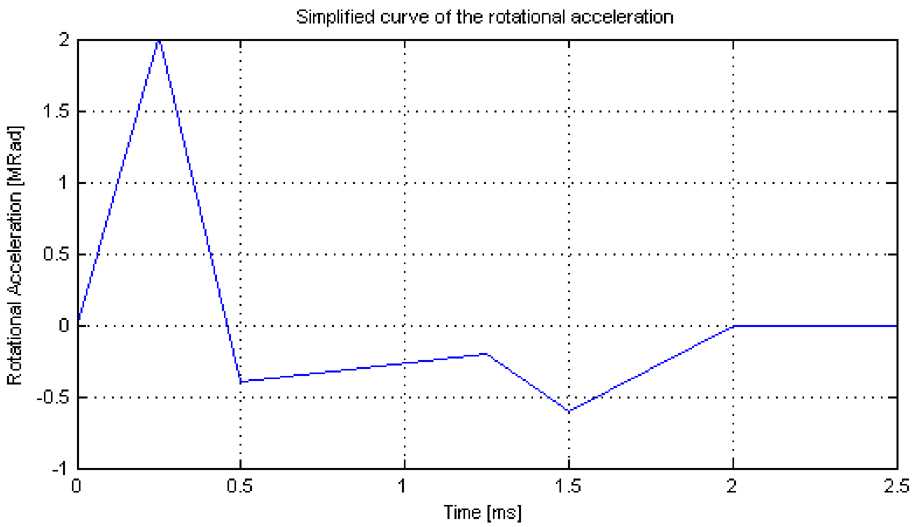

Figure 4) the simplified rotational acceleration load. With the centre placed roughly at the same place in the model as in the experiment.

The brain stem does not extend outside the skull in this model of the rat head. After a comparison with a boundary condition that forced the end of the spine to move with the rotation of the head and without this boundary condition, the boundary conditions that described the displacement of the spine were chosen. Because the fixed movement with the rotation was the most physically sound way to model the movement of the brain stem.

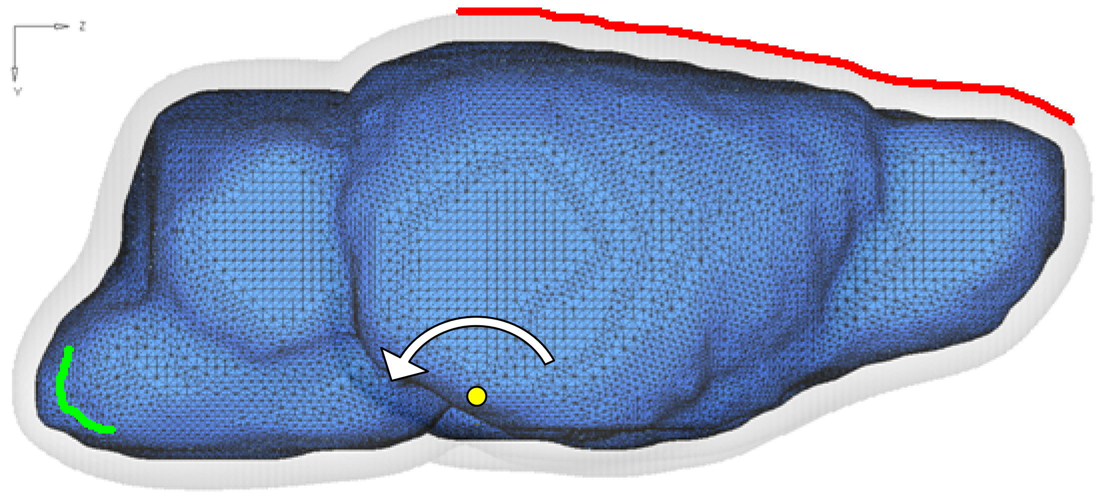

Figure 5) Showing the two boundary conditions that are used during the simulations. The red line is showing where the boundary condition on the skull bone is applied; the green line is sowing where the boundary condition on the spine is applied. The yellow dot shows the centre of rotation and the arrow shows the direction of the rotation. The transparent layer is the skull bone and the blue mesh is the CSF layer.

3.1.2 Modelling the skull bone

In the mesh provided by *Gianfranco* (2010) the skull bone was modelled as one layer of shell elements. Since the load is applied on the outside of the skull bone the shell layer could introduce errors to the model. When using a shell layer the applied load could just as well been applied to the inside of the skull, which is obviously not true. A question was also if the skull should be assumed to be rigid i.e. investigate how big the deformations will be during the loading. Two different ways of modelling the skull bone was investigated, a layer of shell elements and three layers of solids.

3.1.3 Comparing tetrahedral and mixed mesh

Two different models were compared, where the only difference was the element types in the mesh. One model comprised of tetra-elements only and the other comprised of a mix of tetrahedral elements and hexahedral elements. As a consequence of the different element types the number of elements differed in the two models. For the model with mixed elements there was approximately 900 000 solid elements and for the model with only tetrahedral element there were 3 200 000 element. All other aspects of the model were kept constant.

3.1.4 Centre of rotation

The centre of rotation was first estimated using information from *Davidsson et al* (2009). The height of the centre of rotation was then varied to assess the sensitivity to centre of rotation on the results. The chosen variation interval correlates to the natural difference in height of the rat skulls. The reason for these variations were that the size of the head of the rats was different from one subject to another and the top of the skull bones were in all experiments mounted to the test rig on a fixed distance from the centre of rotation.

3.1.5 Sensitivity test of the shear stiffness

To investigate how sensitive the model was with respect to the stiffness of the brain material, a sensitivity test was conducted. Where two general cases where the long and short term shear modules are doubled and halved. These two cases were called hard and soft since they are harder and softer then the material given in the literature.

3.1.6 Local stiffening

In all simulations the brain has been assigned the same bulk modulus value, i.e. *2.19* [GPa]. The bulk modulus describes the substance's resistance to uniform compression, and this value seen in brain material is almost the same as for water. The effect of a harder brain part (Corpus Callusom) is studied. The bulk modulus is doubled to show the effects.

3.2 Data processing and analysis

The different types of data that were used for the comparisons of the different simulations were: strain, shear stress and von Mises stress. The data were not filtered and were presented in its raw format. Stress was calculated from the deformation of the element. The Pressure was calculated from the change in volume of the Elements.

As described in *Section 1.1.1* a number of different properties and their influences during the simulations will be investigate. The evaluation will consist of a visual comparison between different simulations to see how they behave for different conditions. Cross sections of the model were provided at the same place in the FE model as in the analysis of the tissue damages following the experiments with the help of three nodes; with these three nodes new planes were created that represent the sections used in the injury assessments.

A slice along the brain was also created to investigate other parts of the model. The slices are compared at the same time step in the simulation for an objective comparison.

The maximum values of strain and stress in the model could not be used to analyse the behaviour due to the numerical errors in some of the tetrahedral elements inside the model.

4 Simulation results

A number of different simulations were carried out to develop an injury mechanism that can predict the injuries observed in the experiments in the rat brain. In addition, simulations were done to evaluate the numerical behaviour of the model.

4.1 Different methods for modelling the skull bone

Two different methods were considered i.e. the skull bone as a layer of shell elements and finally three layers of solids. In both cases the thickness of the skull bone was assumed to be one millimetre. As can be seen in *Figure 6*, there were at least four different ways to model the one millimetre thick skull bone. It can also be seen that when using three layers of solids the elements were cubic, which was most appropriate for the numerical solution.

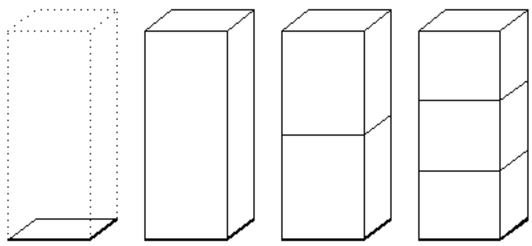

Figure 6) Schematic representation of the possible element shapes that could be used to model the skull bone. The element to the left is a shell with the thickness represented as the dotted lines. The following elements are the one, two and three layer solids. Only the shell and three solid layer elements was used in this simulation. In the future simulations the three layer solid elements are used.

The average side of an element in the model is *0.3* [mm] and the thickness of the skull is approximated to be *1* [mm], the thickness of the shell were greater than the length of the side. It could then be questioned if the shell formulation is appropriate. The advantage of a shell formulation is the computation time. One or two layers of solid elements were not used because the solid elements should be as cubic as possible; which they are if three layers are used.

The highest stress in the elements for two different methods are plotted, in *Figure 7*, a dampening effect can be seen with the skull modelled with tree layers of solid elements.

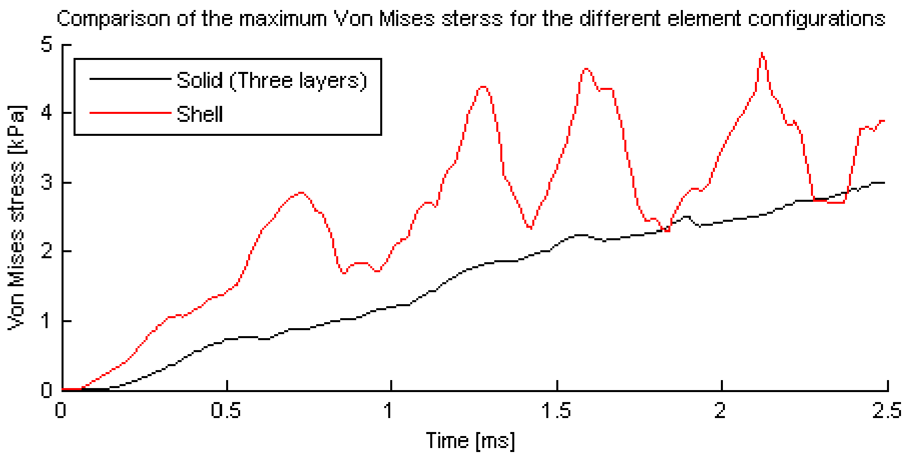

Figure 7) Maximum von Mises stress found in the brain tissue with a layer of shell elements and with three layers of solid elements. Showing a dampening effect of three layers of solid elements.

Visible differences between the two methods were found during the simulations of the brain. The difference found in the simulation of the rat brain is a result of the dampening that is introduced by the solid elements. Therefore a rigid skull would not be good, since a rigid skull would not damp the load on the brain or transfer the load to the brain surface correctly.

4.2 Comparing tetrahedral and mixed mesh

Since under-integration is used in both models, i.e. the model with only tetrahedral elements and the model with both hexahedral and tetrahedral elements, the number of elements is the same as number of equations to be solved for each time step. But the number of elements in the tetrahedral model is three times as many.

The computation time for the mixed mesh is 30 h and the computation time for the tetrahedral mesh is almost 215 h on a computer node with four 3 GHz cores running LS-Dyna 971 in one of the Linux clusters at Chalmers.

4.2.1 The strain comparison

The logarithmic strain can be seen in *Figure 8*. When the strain values are compared, a clear difference in the two different, i.e. the tetrahedral model and the mixed model, models can be seen, where the tetrahedral model indicates numerical errors and the mixed model shows a reasonable response to the load, both in strain values and where the strains occur.

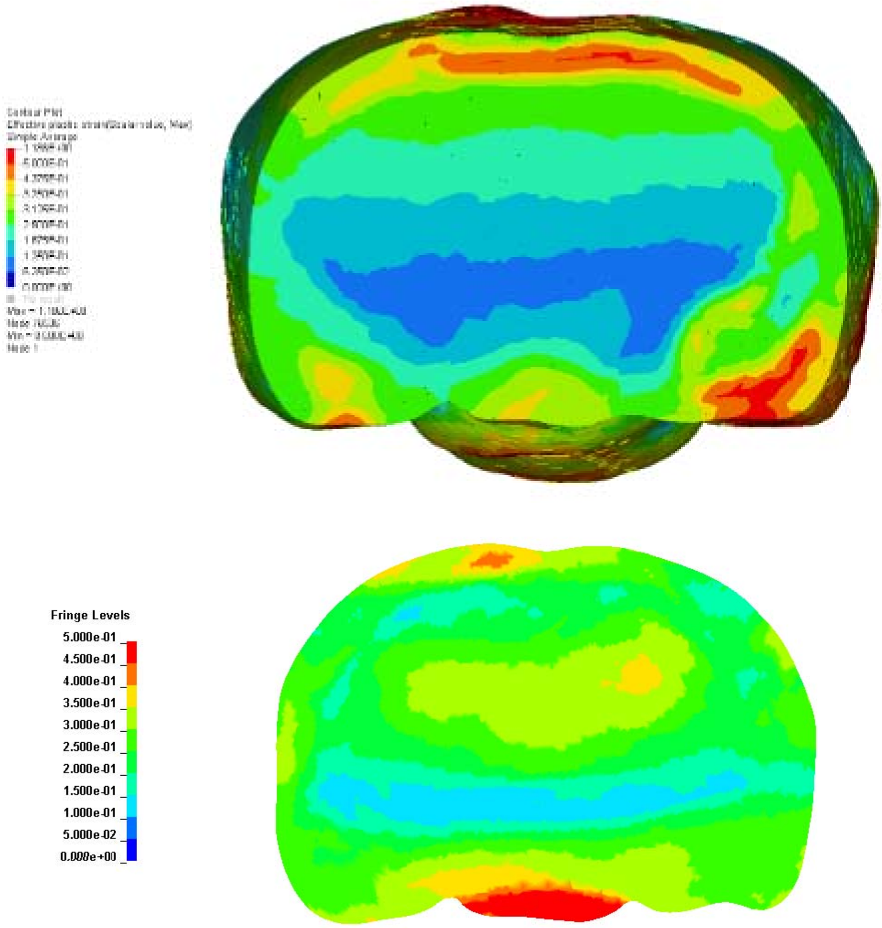

Figure 8) The strain for the two different meshes, The mixed model on the top and the tetrahedral mesh below. The strain is for the last time step in the simulation. The Axis of the logarithmic strain are set to vary from 0 – 50% for both cases. For both cases it is hard to see a good correlation to the injuries seen in Figure 1.

The strain is highest in the top of the brain in the mixed mesh (60%) and on one of the lower sides in the tetrahedral mesh (70%). The threshold level, found in the literature, for injuries to the axons are 20 % and above. For strain levels 30% – 60 % vascular ruptures are expected in bridge veins. *Kleiven* (2007)

The strain is located in roughly the expected regions, for the mixed model. The level of strain is also in the expected range, as given by *Kleiven* (2007), or slightly above. For the model with only tetrahedral element it is hard to see any correlation with the injury, it is therefore hard to draw any conclusions on the correlation between the strain and injury.

4.2.2 The shear stress comparison

The shear stress is not good for the brain, since this might cut the material. It can be reasoned that the axons are more sensitive to the shear than to other stress, since they are long and thin. A comparison of the shear stress can be seen in *Figure 9.*

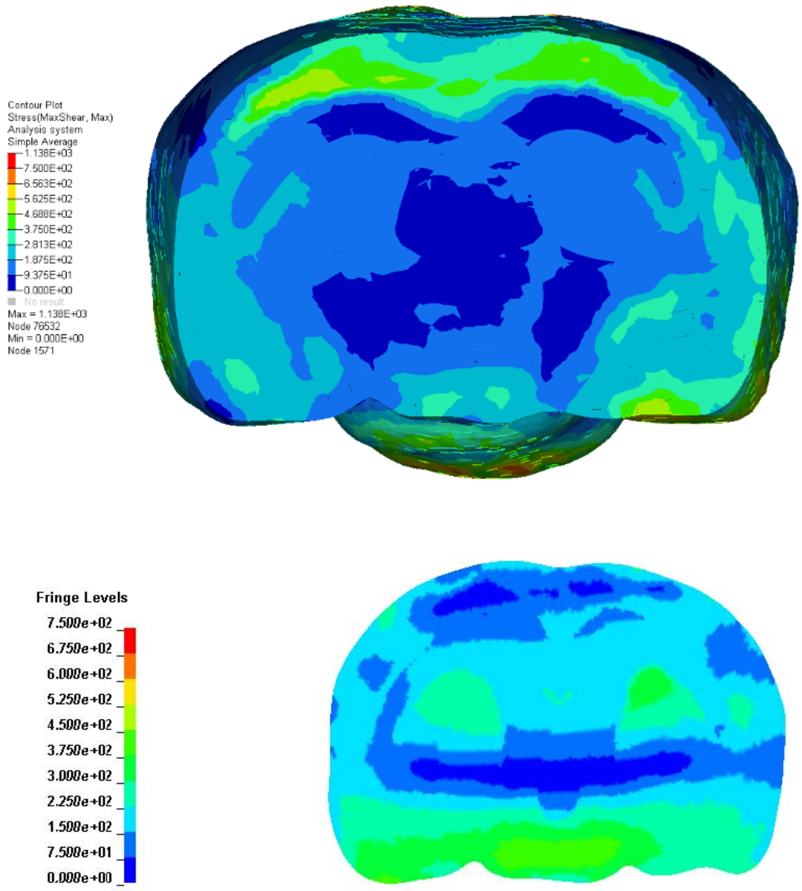

Figure 9) The shear stress for the two different meshes, the mixed mesh on the top and the tetrahedral mesh below. The concentration of shear has a good correlation to the injuries seen in the experiment for the mixed mesh, and rather poor for the tetrahedral mesh. The range in both models is 0 -750 [Pa]. The highest levels can be found

The tetrahedral mesh gave a pore correlation to the injury data. However, the mixed mesh did correlate well to the injury seen in the brain. The maximum shear stress in the section was approximately 500 [Pa] and the value was found at the location of the injury.

The location of the shear indicates that there is a connection between injury and shear stress. Tough from the given data it is hard to give an exact threshold for the injury. The values are related to material data that is used in the simulation, but they can be used as an indication of where injuries are likely to arise.

4.2.3 The von misses stress comparison

The von Mises stress is an average of stresses and therefore it is not surprising that we see similarities to the shear stresses. This can be seen when comparing *Figure 9* and *Figure 10*. The von Mises stress can be seen in *Figure 10*, and does not differ much from the shear in *Figure 9*.

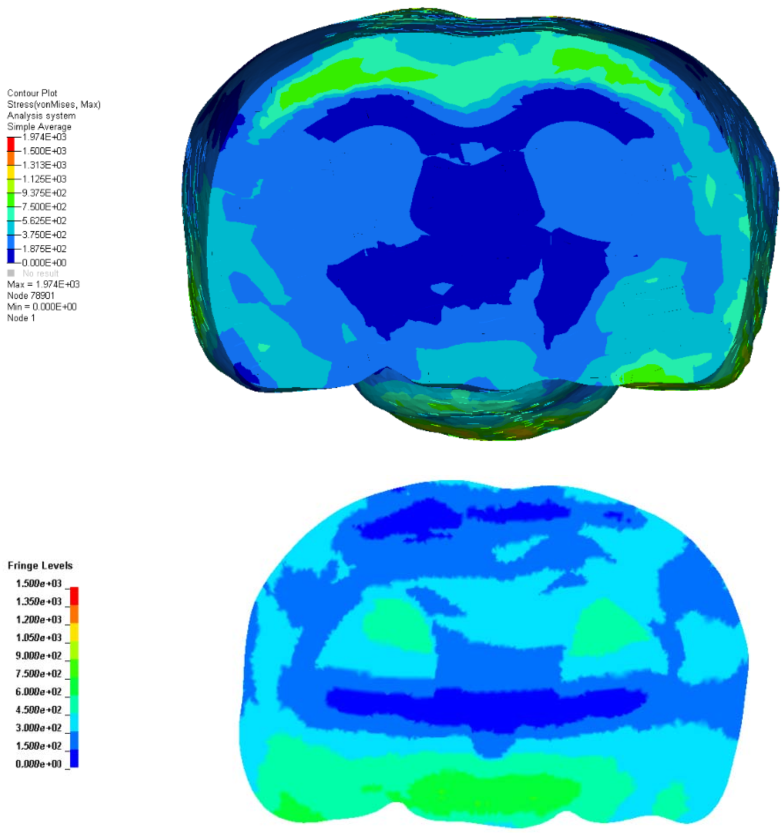

Figure 10) The Von Mises stress at half the simulation time i.e. after 1.25 [ms]. The range is 0 – 1500 [Pa] and the highest value seen in this cut is approximately 1 [kPa]. The mixed mesh on the top and the tetrahedral mesh below. The maximum stress is approximately 750 [Pa] in the section.

Since concentrations in shear stress and von mises stresses can be seen at the same location it is hard to tell which stress is the governing mechanism behind the injury. But as states above it is probably the shear stress. It is worth noting is that the von mises stress is approximately twice as high as the shear. However, the relatively high values of von mises stress should not be used as a single indication of injury; as there may be a big difference in how well the material is able to endure shear and compression stress.

4.2.4 Concluding remarks on the comparison of the two models

When looking at the results, it can easily be see that tetrahedral elements should not be used to mesh the brain. The reason to why the tetrahedral mesh does not correlate well to the experiment is due to the fact that tetrahedral elements are stiffer than cubic elements. Because brain mater is such a soft material the numerical stiffness introduced by the tetrahedral elements will then govern the solution for the tetrahedral mesh. Therefore the tetrahedral model will be invalid. This could have been seen from the Hourglass energy, since the hourglass energy is close to zero for the tetrahedral mesh. For the mixed mesh the hourglass was 5% of the total energy, which doesn't stiffen the model more than necessary.

It is hard to give a threshold for when the injuries starts to occur. However, it looks like the von Mises stress and shear stress correlates with the injuries between the cortex and the corpus callosum. It could be investigated future where for what injuries the tissue ruptures.

The strain level is 30% - 50% at the top of the brain in the mixed mesh simulation. This strain level is the same level as the braking of human cerebral veins and arteries rupture, *Kleiven* (2007). *Takhounts et al* (2003) say that the strain level where injuries occur is 40% - 70%.

4.3 Effects of a variation of the centre of rotation

During the experiment done by *Davidsson et al* (2009) variations in the centre of rotation occurred due to the way the skull was attached to the experiment apparatus. From this it was investigated if this small change in centre of rotation would have a non negotiable effect of the simulation.

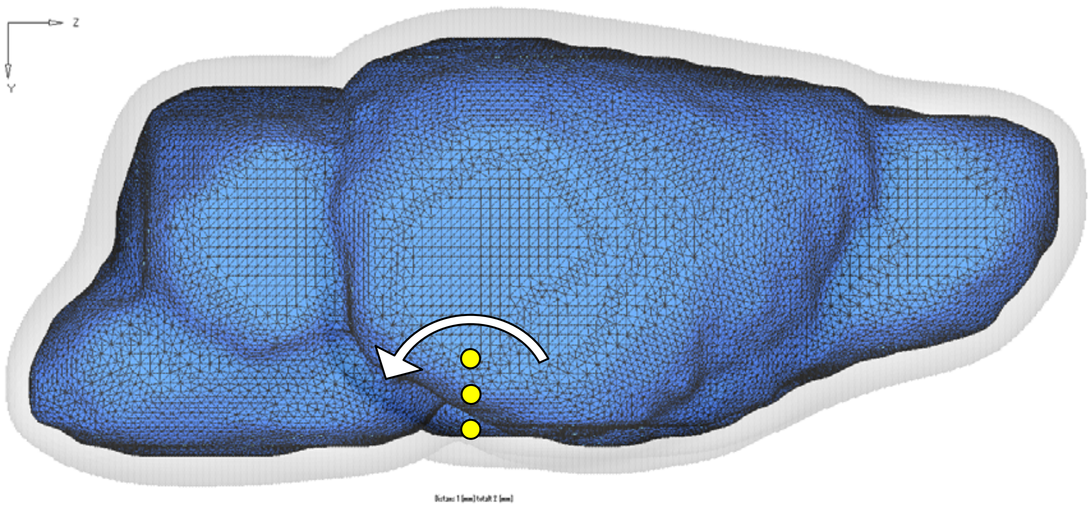

Figure 11) the three different centre of rotation the distance between the rotation centres are 1 mm, i.e. two millimetres from the highest to the lowest centre of rotation. This is a attempt to reproduce the natural variation in the experiment, where the rats has different heights witch in its turn results in a centre of rotation that is placed at different heights. In the figure the blue part is CSF, and the transparent part is the skull bone.

The results from the simulations are shown in *Figure 12* and *Figure 13*. It can easily be seen that the small variation does not affect the results to a great extent.

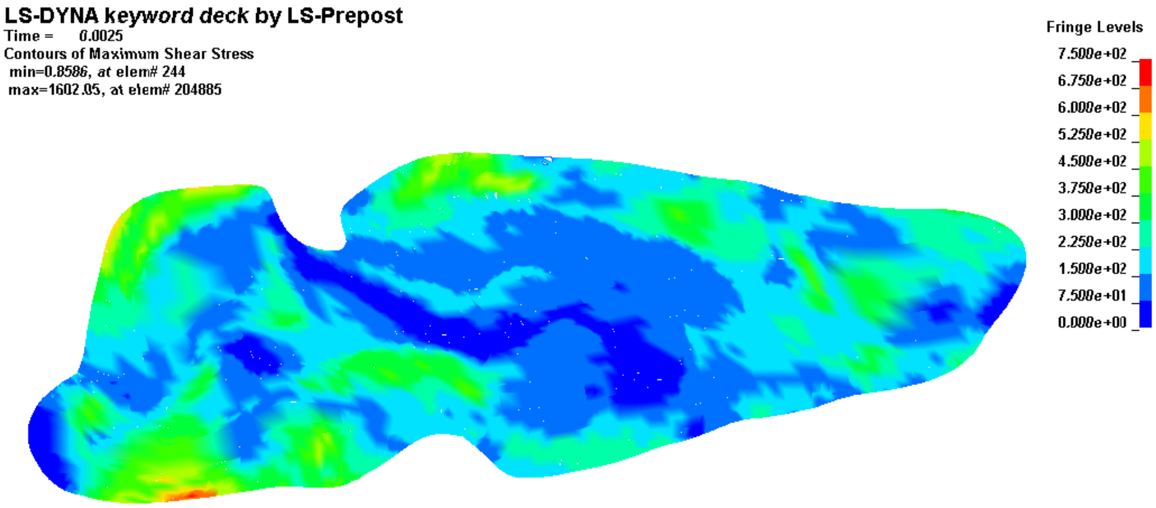

Figure 12) plot of the maximum shear in the last time step when the centre of rotation is shifted up 1 [mm, a cut taken in the centre of the brain.

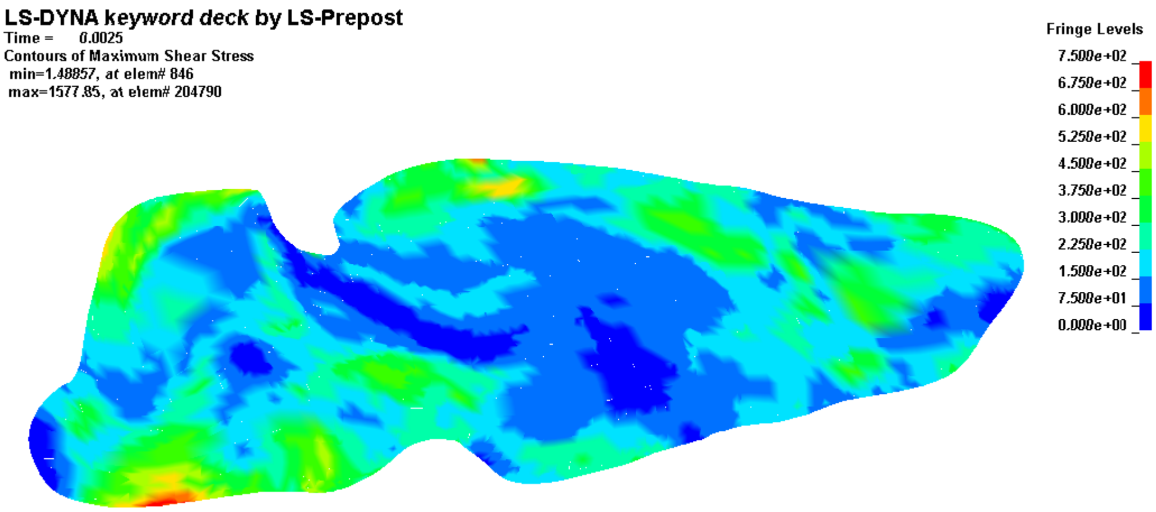

Figure 13) plot of the maximum shear in the last time step when the centre of rotation is shifted down 1 [mm], a cut taken in the centre of the brain.

The pressure comparison as can be seen in *Figure 14*. The pressure response of the two cases differs with the change in centre of rotation, since pressure can be seen as a function off acceleration and density of the material in the brain this is not strange. But this shows that the centre of rotation need to be well defined for a validation of with pressure.

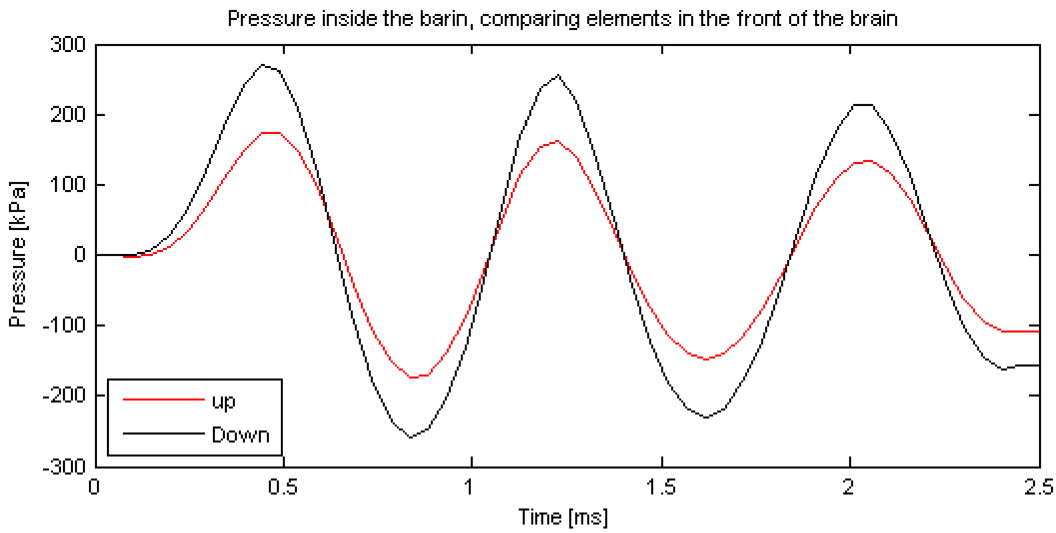

Figure 14) When moving the centre of rotation up and down one millimetre form the original position, taken from the front of the rat brain. The pressure for the original position is not shown, but is in between the two cases.

Form the variation of the centre of rotation one can conclude that small variation in centre of rotation has effects on the stresses in the brain. But these effects are too small to draw any conclusions regarding the magnitude of the injury.

4.4 Sensitivity test of the shear stiffness

As seen in *Figure 15* and *Figure 16* there is an obvious difference in von Mises stress between simulations using the soft and the hard brain tissue representations. The von Mises stress is higher in the hard case, which is not strange since the stiffness is increased.

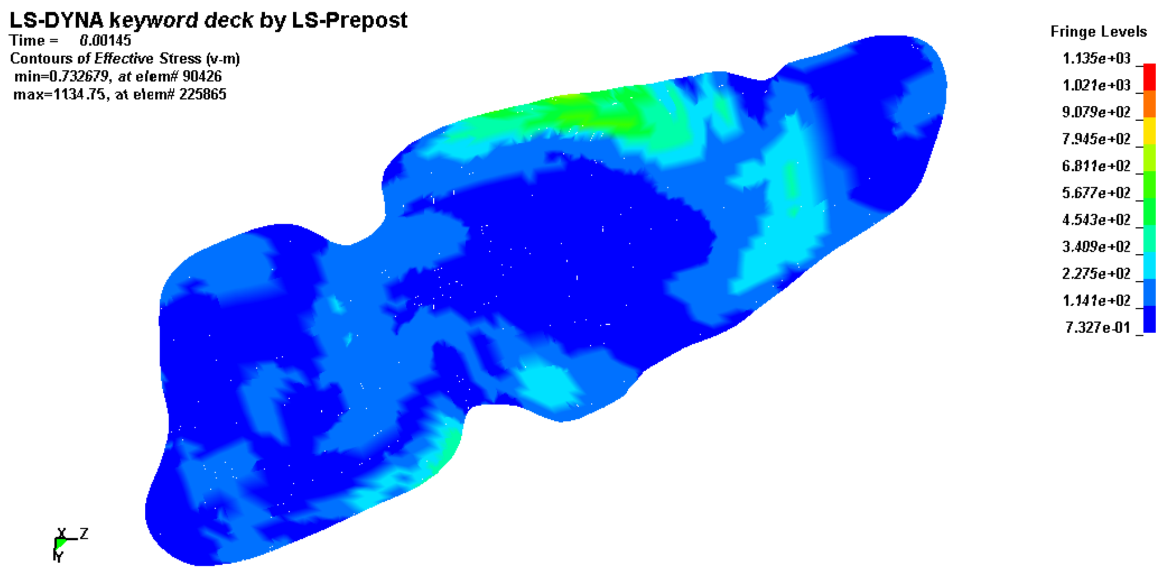

Figure 15) Soft, the shear modulus are half of the one applied to the original model. The maximum von mises stress seen in this cut is 800 [Pa], 1.25 [ms] into the simulation.

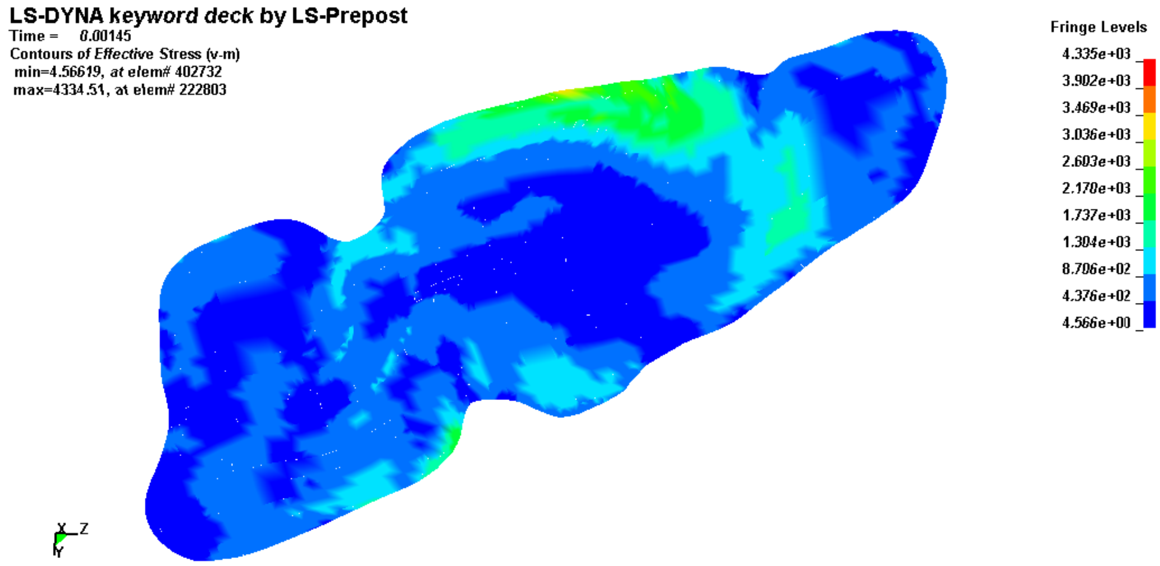

Figure 16) Hard the shear modules are twice as high as in the original model. The maximum von mises stress seen in this cut is 3000 [Pa], 1.25 [ms] into the simulation.

When the two different cases are compared, the same general shape of von Mises stress can be seen. Differences in the amplitude of von Mises stress can be observed but the same shapes can be seen. There are also additional stress concentrations seen in the harder model.

And as can be seen in the pressure plot in *Figure 17* there is no significant difference in pressure between a hard or soft model.

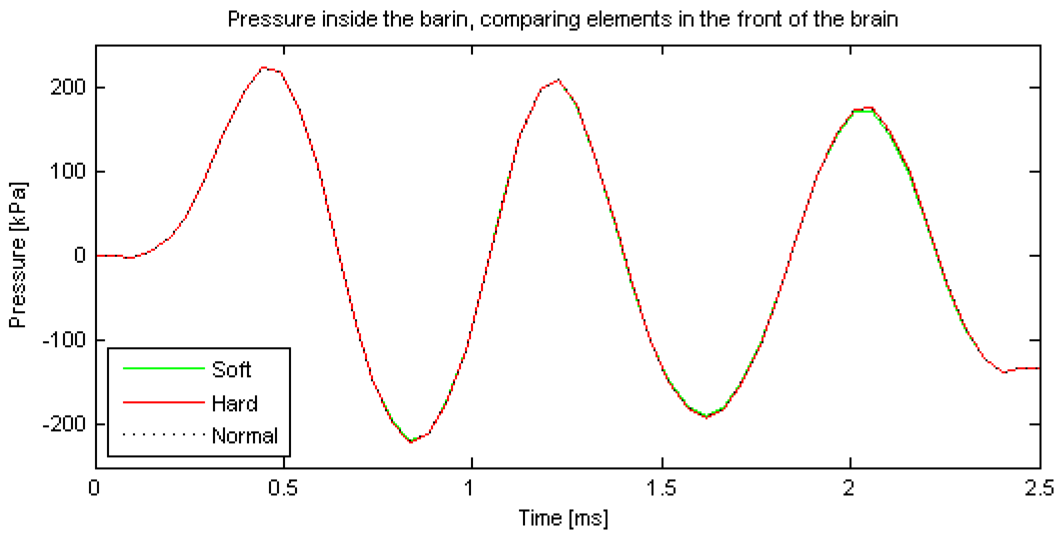

Figure 17) The pressure extracted from a element in the front of the brain. As can be seen there is no significant difference between the cases.

The last comparison between the soft and hard case is the change of displacement due to the change in stiffness. For this comparison the position of a node was extracted for all time steps in the simulations. The resulting path of the movement can be seen in *Figure 18*. The largest difference was seen in the end of the simulation where the position of the nodes differs 0.1 mm in x-direction. This small change in deformation will not easily be measured during the experiment.

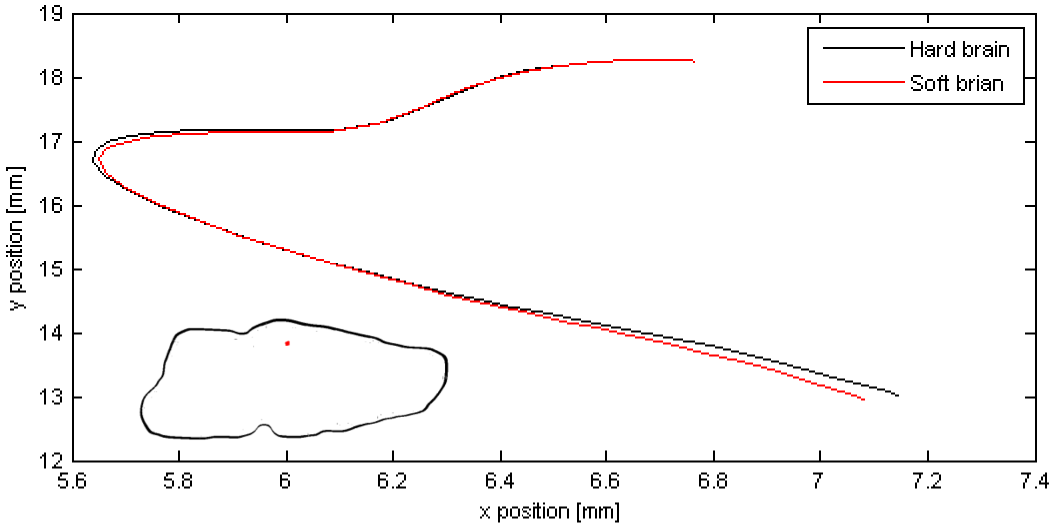

Figure 18) The trace of the same node for the two different stiffness's. The position of the node can be seen in the left corner of the figure.

4.5 Effects of a locally increasing the stiffness

When the tetrahedral and mixed mesh was compared it was reasoned that the difference came from a numerical stiffness, introduced by the tetrahedral elements. Therefore it was investigated how the model would be affected if one part was stiffer then assigned. Therefore the stiffness of the corpus callosum, was doubled.

In the comparison of the mixed and tetrahedral mesh it could be seen that the shear stress and stain was concentrated to the interface between the corpus callosum and the cortex; where the outlines of the corpus callusom can be seen clearly in *Figure 19*. However, when the corpus callusom was stiffened, the stress and strain moved its place and magnitude.

Corpus callusom was selected since this part is close to where the interesting results are found. And therefore it was reasoned that it would be of interest to see the behavior of the model when the stiffness was modified.

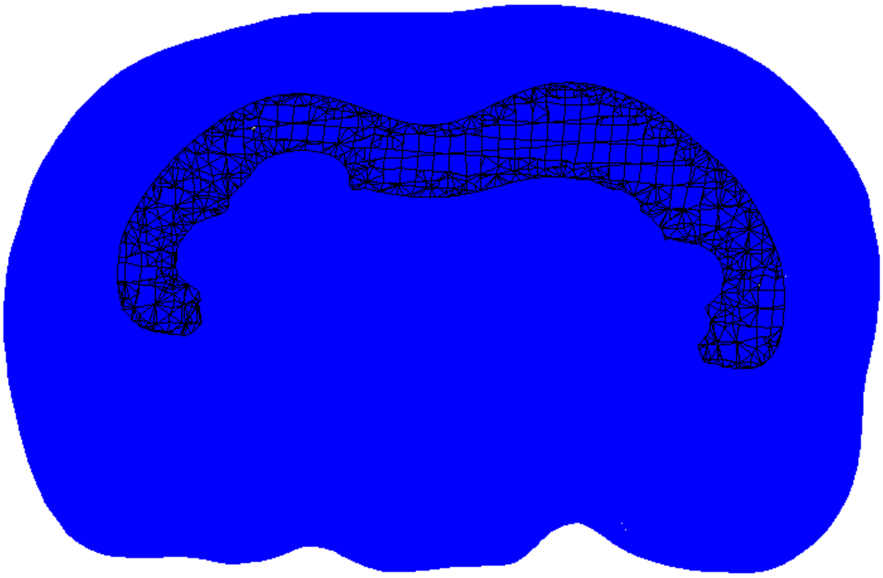

Figure 19) Mesh of the corpus callosum, inside the brain. Where the rest of the mesh is suppressed. The cortex is placed above the corpus callosum.

When looking at *Figure 20* the strain can be seen. The strain was higher than in the original model but was more or less in the same place as in the first model. The strain levels are higher over the whole cross-section.

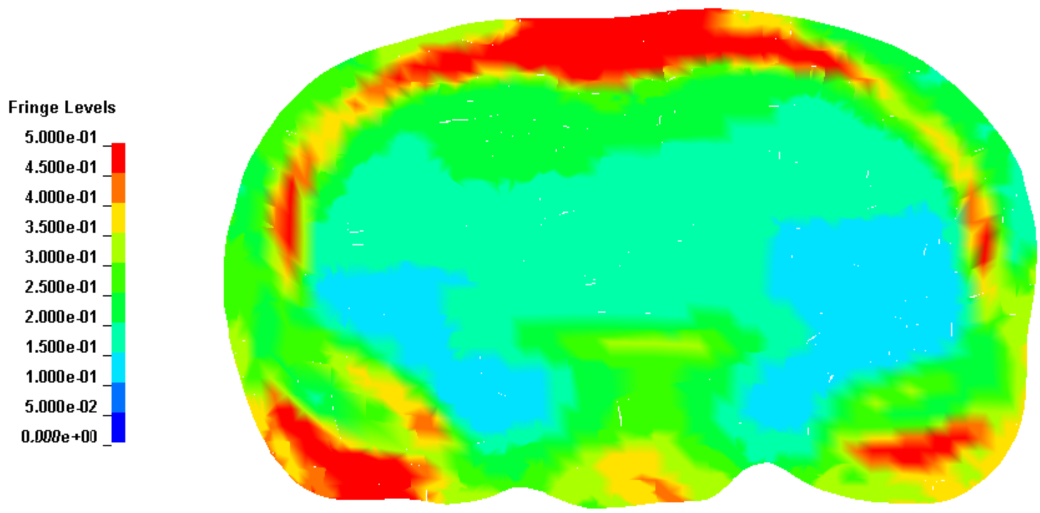

Figure 20) The strain in the mixed model with a stiffer corpus callusom. The highest value found in this plane was approximately 80 % but for easy comparison the same strain range was used as previously i.e. 0 – 50 %

When the shear stress seen in *Figure 21* was compared to the original model it can be concluded that the magnitude of the shear stress not change much. However, the concentration of shear was moved down the sides of the brain, this does not correlate with the injury.

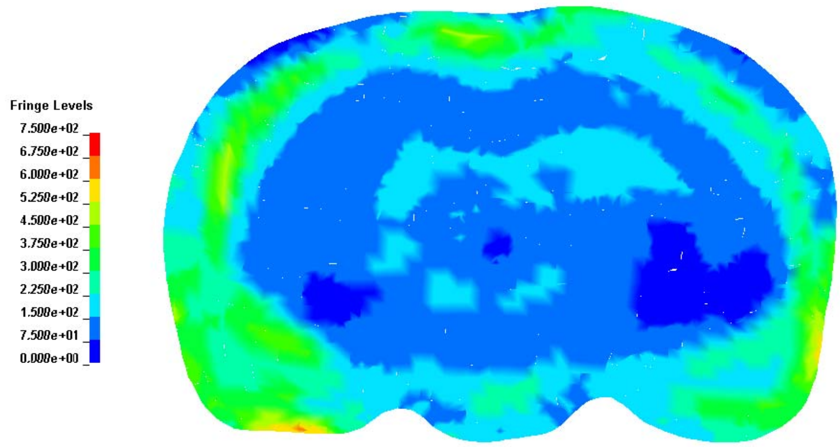

Figure 21) Shear stress for the same plane as previously used. The range was set to 0 – 750 [Pa]. However, the maximum value seen is around 600 [Pa]. The concentration of shear stresses has moved down to the sides of the brain with stiffer a corpus callusom.

It can be concluded that the right stiffness of pars of the brain need is needed to get good results. Therefore the simulation with only tetrahedral elements did not give a good representation of the experiment. The increase in the stiffness of the corpus callusom was done to show the affect of the wrong stiffness. But there is data that suggest that the stiffness in the mixed model is too low.

5 Discussion and conclusions

The most interesting and promising results in this thesis is the correlation of the injury and the shear stress. Stress and injury correlate surprisingly well. An absolute threshold for when the shear stress starts to inflict an injury is not given form this study.

5.1 The model

It was soon discovered that the use of different element types introduce free edges in the mesh. This would according to *Engineering Research AB (ERAB)* make the structure weaker than prescribed. The number of tetrahedral element was high compared to the hexahedral element and the geometry of the brain is complex, therefore it was impossible to convert the existing mesh to a version that only includes hexahedral elements with the given software and with the given time span. However, a mesh with only tetrahedral elements was generated that could be compared to the mixed mesh.

Because of the incompressibility of the soft tissue used to describe the brain tetrahedral elements is not a good choice for the simulation. Tetrahedral elements will make the model more sensitive to volume locking. *Joldes et al* (2008)

When tetrahedral and hexahedral elements are mixed a free edge will always be present on the tetrahedral element. This problem can be seen in *Figure 22* and will lead to a discontinuity in the mesh during deformation. This mismatch will always be present if the different elements are mixed, explained by *Ottosen and Petersson (1992*).

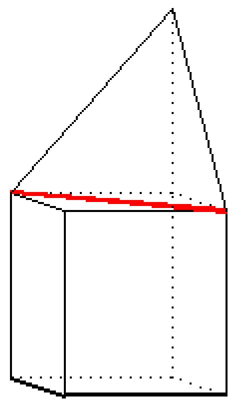

Figure 22) The figure shows two different element types, the tetrahedralelement on top of a hex element. The red line indicates the edge that is not compatible between the two element types.

The geometry that describes the rat brain could be improved. It will be hard to improve the mesh with the software and images currently available at Chalmers. The mesh should be fine enough to capture the important features of the brain, while at the same time still simple enough to easily be analyzed. The mesh creation is of high importance since this will have a large impact on the outcome on the model where the mesh is used.

The model behaviour can be considered stable, but a number of elements were deleted during a simulation. The eroded elements did not affect the solution to a great extent. But this could be addressed with the review of the mesh and solved with the new elements.

When a tetrahedral mesh was used, with approximately the same number of elements as for the mixed mesh, the results were worse than for the one used in the final comparison. The solution became instable, which was first believed to be due to a small error in the axis as the brain rotates around, the comparison done now indicates that it is likely that this was related to the stiffness of the tetrahedral element. Tetrahedral elements have been used to model an impact of a liver, *Bourdin* (2010), and it was concluded that the tetrahedral element would suffice. However, my findings of the rotating rat brain indicate this assumption is inappropriate.

5.2 Strain

The strain did not correspond as well to the injury as was expected. Since the axons are long, the axons have a flexibility to stretch, however a focal stretch can occur and give an injury to the axon *(Gentleman et al 1995)*. But such phenomenon could not be captured in these simulations.

5.3 CSF

In this model the layer of CSF surrounding the brain was modelled whit an incorrect thickness. The volume of CSF inside a human brain varies with age and for an adult human the CSF is 170 [ml] where the total weight of a brain is roughly 1500 [g] according to *Pollard et al* (2004).

Where the average rat brain is roughly 2 [g] the CSF layer of the rat should be in the order of 0.25 [ml]. However, the volume of the CSF in *Gianfrancos (2010)* model is 1 [ml], if the same ratio between the weight and CSF volume is the same, the volume of the CSF layer is almost four times too great. This will affect the model and make the response softer than expected since the CSF has a dampening effect. However, the thickness of the CSF layer in a rat might be too small to model with solid elements and the use of a solid fluid interface is a better solution to this problem.

5.4 Pressure

The pressure has been problematically during the work of this thesis; the problem is related to the mesh that was given in the beginning of the work. The pressure response looks poorest in the transition regions between different parts. This is no surprise since the highest concentrations of tetrahedral elements can be found in these regions and the tetrahedral elements are known to be unstable when calculating the pressure.

When looking at the result of this thesis the problem with the pressure should be kept in mind, and therefore future studies are needed. The problem with the pressure might be isolated to the pressure alone, but this is highly unlikely. When changing the elements from tetrahedral to hexahedral elements the results might change, however it is hard to guess how big the impact will be on the stresses and strains. Therefore it is hard to give any final conclusions regarding the injuries on the rat brain.

Pressure is often used to validate models of the brain therefore it is of interest to solve this problem. Future study is needed to address the problem with the pressure. When the pressure problem is solved the effect of the tetrahedral element can be evaluated.

It is possible to solve the pressure problem in LS-Dyna 971, however not necessarily with the given mesh.

From the pressure comparison in the sensitivity test it can be seen that even if the pressure is validated it does not necessarily validate the stiffness of the material. However, the pressure could be used as guidance for the validation of the centre of rotation, rotational acceleration and density in the model.

6 Recommendations

Due to the high ratio of tetrahedral elements in this mesh it is highly recommended to review the current mesh and remove all or as many as possibly tetrahedral elements. When doing crash simulations it is of interest to minimize the number of tetrahedral elements due to their properties described in the discussion. The software used to generate the mesh does not have any good way of controlling the generation of tetrahedral or hexahedral elements, i.e. only chose mixed mesh or tetras only can be chosen. Therefore it should be investigated if there is any other software which provides a improved control over the mesh generation.

A comparison of the simulation results to the real experiment pressure data from the experiment can be done to validate the pressure response; but this does not necessarily validates the strain and stress response. Due to the current state of the model the pressure cannot be acquired from an arbitrary point in the brain; which would be desirable.

To validate the deformation of the brain tissue a high speed x-ray apparatus can be used to investigate the movement inside the brain. A problem with this approach is the requirements of the high speed x-ray apparatus needed. The experiment is carried out for roughly *2.5* [ms] including the on-loading and the off-loading phases. At 200 Hz the exposure time between two frames is 5 milliseconds. This could be used to trace a marker during the simulation. However the size of the rat brain is a problem, since the neutral density markers should be small compared to the brain. And the markers should be inserted without damage the tissue. If using an x-ray apparatus with an exposure time greater than *2.5* [ms], the trace of the markers can be studied. But the intermitted positions cannot be observed. The trace can then be compared to the corresponding node, as done by *Takhounts et al (2003).* With a higher frame rate the intermediate position of the trace markers could be extracted.

It's hard to guess the quality of the images that can be extracted from a high speed x-ray apparatus, but the displacement inside the brain should be possible to extract. However, the stiffness of the brain can be hard to determine since the difference in the deformation seen in the sensitivity test was small; the difference of the movement was less than 0.1 [mm] as can be seen in *Figure 18*. This would suggest that the small size of the brain makes it impossible to determine the stiffness of the brain with an x-ray apparatus. However, it can validate the boundary conditions and connections of the parts in the brain. To validate the stiffness of the model the key is to validate the deformation in the simulations to the deformation in the experiment. This has been done by invasive methods in the past, with force displacement curves as outputs, *Brands* (2002).

7 References

Bourdin. X, Trosseille. X, Petit. P. (2007): *Comparison of tetrahedral and hexahedral meshes for organ finite element modeling: an application to kidney impact.* Bron, France. 2007

Bradshaw. D. R. S and Morfey C. L. (2001): *Pressure and shear responses in brain injury models.* ISVR, University of Southampton, UK,

Brands W.A. Davy (2002): *Predicting brain mechanics during closed head impact: numerical and constitutive aspects*. Eindhoven, Technische Universiteit Eindhoven (2002)

Davidsson. J, Angeria. M and Risling. M. G (2009): *Injury threshold for sagittal plane rotational induced diffuse axonal injuries*, IRCOBI, 2009, pp 43-57

Joldes R. G., Wittek A., Miller K. (2008):*Suite of finite element algorithms for accurate computation of soft tissue deformation for surgical simulation*, Medical Image Analysis, Vol. 13 (2009), pp. 912-919

Gentleman S. M, Roberts G. W, Gennarelli T. A, Maxwell W. L, Adams J. H, Kerr S, Graham D. I (1995): *Axonal injury: a universal consequence of fatal closed head injury*. Acta Neuropathologica, Charing Cross and Westminster Medical School Department of Psychiatry, London UK

Hallquist. J. O. (2006), *LS-DYNA 971® Theory manual*. Livermore Software Technology Company. Livermore, California USA

Mao, H., Zhang, L., Yang, H. K. and King, A. I. (2006): *Application of a Finite Element Model of the Brain to Study Traumatic Brain Injury Mechanisms in the Rat*. Stapp Car Crash

Hardy N. Warren, Khalil B. Tawfik and King I. Albert (1994): *Literature review of head injury biomechanic*, Bioengineering Center, Wayne State University, 818 W. Hancock, Detroit, MI 48202, U.S.A. 1994.

Kleiven. S (2007): *Predictors for Traumatic Brain Injuries Evaluated through Accident Reconstructions,* Stapp Car Crash Journal, Vol. 51 (October 2007), pp. 81-114

Kleiven, S (2002*): Finite Element Modeling of the Human Head*, Report, Department of Aeronautics Royal Institute of Technology, Stockholm, Sweden, 2002

Lauret, C (2006): *An Overview of In-Vivo, In-Vitro, and Finite Element Models Developed to Study Traumatic Brain Injury.* Department of Biomedical Engineering, Eindhoven University of Technology, Netherlands

Ninds, *(2010): http://www.ninds.nih.gov/disorders/disorder_index.htm.* National Institute of Neurological Disorders and Stroke

Shulyakov. V. A., Fernando F., Cenkowski S. S., Del Bigio R. M. (2009): *Simultaneous determination of mechanical properties and physiologic parameters in living rat brain*. Biomech Model Mechanobiol (2009) volume 8 pp. 415–425

Pollard J. A., Finn A. (2004): *Hot topics in infection and immunity in children*, Volume 2. European Society for Paediatric Infectious Diseases, University of Oxford. Dept. of Paediatrics. UK

Ottosen, N., Petersson, H. (1992): *Introduction to the Finite Element Method*, Prentice Hall, New York

Gianfranco. R. (2010): Thesis, Department of Injure Prevention, Chalmers University of Technology, Gothenburg, Sweden, 2010, Manuscript in preparation.

Takhounts EG, Eppinger RH, Campbell JQ, Tannous RE, Power ED, Shook LS. (2003): *On the Development of the SIMon Finite Element Head Model.* Stapp Car Crash Journal. Vol. 47 pp. 107-33

Zou. H. (2007): *Analysis and Modeling of the Biomechanics of Brain Injury under Impact*, Ohio State University, USA