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A Computer Model of the Human Head to Assess Mechanical Brain Loading in Car Collisions

R.S.J.M. Verhoeve

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1. Introduction

1.1 Problem definition

In order to study the effectiveness of passive safety measures for road vehicle occupants, both physical (crash dummies) and mathematical/numerical models (multibody- and Finite Element models) of the human being are currently used. For a better assessment of this effectiveness, these models need further improvement. Besides improvement of these models also some of the currently used injury criteria (parameters used for the assessment of the risk to specific injuries) need to be improved. For the assessment of the risk to head injuries the so-called Head Injury Criterion (HIC) is currently used. This criterion is often criticised because it is not based on up-to-date knowledge on the mechanisms that lead to head injuries. HIC is in fact scalar value that is calculated as a function of the linear resultant acceleration of the head in response to the loads applied to it.

Currently Eindhoven University of Technology (TUE) participates in the European research project ADRIA\(^1\). The general objective of this project is to reduce the number and severity of casualties among vehicle occupants involved in frontal collisions in the European Community. The ADRIA project is divided in four different tasks. One of these tasks concerns head injury assessment. The participation of TUE in the ADRIA project is focussed on this task. The main objectives in this task are the evaluation of the Head Injury Criterion (HIC) and the identification of head injury mechanisms, resulting in recommendations for improved head injury criteria.

In order to meet these objectives a methodology was set-up in which real world, frontal impact car collisions are reconstructed. In such a reconstruction, the loads applied to the head of the victim during the collision will be retrieved, as well as values for HIC. After this has been done, the mechanical response of the victim's head to these loads must be determined. Once this response is known, it can be compared with the head injuries that were found on the victims in reality. In this way links may be identified between specific mechanical head response parameters and specific types of head injury. Such links provide information on the mechanisms that cause these injuries, which is useful for setting up future improved head injury criteria. HIC can be evaluated by means of comparison with the predicted head responses and the real head injuries found on the victims for the different accidents reconstructed.

The mechanical responses to the loads applied to the head are planned to be determined by means of Finite Element (FE) head load simulations. However, for performing these simulations a FE model of the human head is needed. An evaluation study on more recently developed FE head models was planned in order to end up with a FE head model that is suitable for use in these simulations. This report presents the results of this evaluation study and the FE head model as it will be used for the head load simulations in the ADRIA project.

\(^1\) ADRIA = "Advanced crash Dummy Research for Injury Assessment in frontal test conditions".
1.2 Outline of this report

As a first step in the FE head model evaluation process a literature study was performed on the head models that have been developed in the current decade. In this literature study attention was paid to the configurations of the head models, the type of simulations they were used for, and the validation of the biofidelity of their predictions. The results of the literature study are presented and briefly discussed in chapter two of this report.

A more detailed evaluation of three of these FE head models (being two versions of a head model developed at Eindhoven University of Technology (TUE) and a head model developed in the United States at NHTSA\(^2\)) is presented in chapters three and four of this report. The objective of the evaluation of these head models was to provide information on the quality and limitations of the models with respect to brain response prediction. In chapter three the configurations of both head models are presented. In chapter four the head load simulations performed with both models are presented and discussed. This chapter concludes with the selection of the head model that is best suitable for use in the victim head load simulations in the ADRIA project.

Further improvement of the selected head model was considered necessary, before it could be used in the ADRIA project. The modifications made to the head model are presented and evaluated in chapter five. The results of the head model evaluation process are summarised in chapter six. In this chapter also the final configuration of the head model, as it will be used for the victim head load simulations, is presented.

\(^2\) NHTSA = National Highway Transport Safety Administration
2. **FE Modelling of the Human Head: general review**

2.1 Introduction

The main purpose of the development of FE models of the human head is to obtain a better insight in the mechanical response to (impact) loading of the human head. The advantage of FE models compared with for instance discrete or lumped parameter models is that the local mechanical response of the different structures of the head can be computed, e.g. in terms of pressures, stresses and/or strains. Spatial distributions of these field parameters can be visualised via contour plots and transient signals via time-history curves.

When a FE head model has proven to be able to simulate the dynamic head response realistically (via a validation procedure), these local response parameters can provide valuable information for getting a better understanding of head injury mechanisms. For instance, specific brain response parameters resulting from head load simulations (such as shear strain levels or local pressure peaks) could be found to correlate with brain injuries sustained during real accidents. The same type of correlation found in several different cases simulated, could indicate a causal relationship between a specific brain response parameter and a type of brain injury. In the procedure described here a validated FE head model is clearly an indispensable tool. Information obtained on the injury mechanisms should next be used for the development of new injury criteria. Once these injury criteria have been developed the FE head model can be used as a valuable tool for the prediction of injury risks during different types of head loading.

Injury criteria that are developed following the procedure described here, would have a more fundamental basis than the currently used Head Injury Criterion (HIC). HIC is an empirical injury criterion that links global (translational) head accelerations directly to head injury. These head accelerations are more easily measured than the intra-cranial response, but they are also not as directly related to the actual injury mechanisms (figure 2-1).

![Figure 2-1: Schematic view of the process of head loads during automotive accidents resulting in head injury. The scheme shows the empirical character of HIC.](image)
2.2 Finite Element head model development: general methodology

In this section a general procedure for the development of FE head models is presented. The procedure can be divided in three parts: set-up of a mathematical model, transforming it into a numerical model via the Finite Element Method and evaluating the numerical model in order to come to a validated FE head model.

2.2.1 The mathematical model

The first step in the FE model development is to determine which anatomical structures of the head should be incorporated in the head model. For a description of the anatomy of the human head is referred to ADRIA interim report D1, on methodology and accident analysis [ADR97]. The structures to be incorporated should be those that are expected to have a significant influence on the brain response during head loading. Once the structures to be incorporated are selected, the next step is to model these head structures in a realistic manner. Whether a head structure is modelled realistically depends mainly on three aspects:

- the geometry and anthropometry of the structure,
- the constitutive model used for the material behaviour of the structure,
- the interface conditions between the structure and its neighbouring structures.

The mathematical head model in fact consists of a set of partial differential equations (conservation of mass, momentum and moment of momentum) valid on a 3D domain and meeting with a set of boundary conditions. The domain is established for each structure by the geometry and anthropometry of the structure. The constitutive model for each head structure has to be specified to obtain a complete set of equations that can actually be solved. The interface conditions between the different head structures form part of the boundary conditions belonging to the mathematical model. The other boundary conditions are formed by external kinematic and/or dynamic boundary conditions, which are specific for the type of head loading.

Data concerning the anthropometry of the complete head and the geometry of the different head structures can be derived from CT/MRI-scans of the human head and/or an anatomical atlas. The constitutive models and belonging material parameter values to be used for the different head structures should be obtained from experiments on biological tissue from the different head structures. Especially for brain tissue it is known that the constitutive behaviour is quite complex (inhomogeneous, anisotropic and physically non-linear behaviour). However, the experimental results that can be found in literature are generally fitted to linear (visco-)elastic material models. Probably the most important interface to be modelled in a Finite Element head model is the skull-brain interface, which in reality is a quite complex interface with three meninges having cerebro-spinal fluid (CSF) between them and blood vessels crossing them. In a FE head model contact definitions for this interface vary from tied interfaces to frictionless sliding interfaces.

2.2.2 The numerical model

After the mathematical head model has been developed a solution procedure for the set of differential equations should be set up. Due to the complex nature of the geometry, constitutive behaviour, interface conditions and the external boundary conditions (including the head loading conditions) no analytical solution is available for the mathematical model. For this reason the model is transformed to a numerical model by using the Finite Element Method [BAT96]. This implies that the continuous domain, determined by the head model geometry, is discretised by dividing it in a finite number of spatial elements connected by discrete nodal points. The manner of dividing the different structures of the head model in elements (meshing) is of significant influence on the eventual quality of the FE head model. Particularly the level of mesh refinement and the quality of the shape of the elements (e.g. aspect ratio's, Jacobian values) plays a role here. Also the choice of type, order and integration method of the
elements are important. In practice most FE head models consist mainly of first order, eight node solid elements and four and eight node shell elements.

Besides the meshing of the model the time integration method used in the FE software is of importance. Most FE head models are developed for FE codes using an explicit time integration method. Although these methods are not unconditionally stable (limitations to time step size), they have a big advantage in comparison to implicit time integration methods in the form of much lower CPU costs. This advantage becomes more important as the FE head models become more complex.

2.2.3 Evaluation of the FE head model

After the numerical model has been generated for a specific FE code, the next step is the evaluation of the head model. This means testing the model on its dynamic behaviour in relation to the real human head. Only when the behaviour of the head model is proven to be in accordance with the reality of the human head the model can be considered validated.

Firstly it is important to check whether the inertial properties (i.e. mass, centre of gravity, moments of inertia) of the model are similar to those of the human head. Whether this is the case depends on the prescribed geometry and mass densities of all the structures that are modelled and of course also on which head structures are not incorporated in the model.

Secondly the head dynamic mechanical behaviour of the head structures should be evaluated. This should be done by comparison of the head load simulation results with experimental data. Two different types of experiments could be used in a head model evaluation procedure: experimental modal analysis of human cadaver head and/or head parts, and sled tests or head impact tests using complete human cadavers or animals. With the first type of experiments the natural frequencies and the belonging modes of the skull, brain or complete head should be determined. A numerical modal analysis with (parts of) the head model should then give information on the biofidelity of the modal behaviour of the model. The sled or head impact tests should give information on the transient behaviour of the different head structures in terms local pressures and/or strains. Human cadaver tests have the advantage of involving human biological tissue, but also have a disadvantage in the fact that in vivo circumstances can only be approximated to a certain extent in this kind of experiments. For this reason in vivo animal experiments are also valuable in a human head model evaluation procedure.

In practice very few experimental data are available that comprise information on local intracranial response to head loading. Head impact tests on cadavers during which intracranial pressures were measured, have been performed by Nahum et al. [NAH77] (see also annex A of this report). One of these experiments is very often used in evaluation procedures for recently developed models. In general it can be stated that there is a lack of experimental data suitable for thorough evaluation of the transient dynamical behaviour of FE head models on a local level.

2.3 3D FE head modelling in the current decade: literature review

2.3.1 General

Over the last three decades several FE models of the head have been developed. The first geometrically realistic finite element model of the human head was presented in 1971, by Hardy and Marcal [HAR71]. This model was used for static simulations, like the FE head model that was presented in 1975 by Ward [WAR75]. Another early FE head model was presented by Shugar in 1977 [SHU77], who used his model for impact simulations. In the following twenty years until present, the FE models developed show an increase in complexity. The FE head models that have been developed between 1971 and 1993 have been reviewed in literature by Ward [WAR82], Khalil & Viano [KHA82] and Sauren &
Claessens [SAU93]. In this section only the recently developed FE head models will be studied. A brief review is given of the set-up of the head modelling studies and the kind of results that were obtained. Specifications of the head models, such as the structures incorporated, element types, the material models used and the material parameters chosen, are presented in annex A.

2.3.2 Wayne State University

2.3.2.1 Ruan's FE head model (WSU model I)
In 1993, Ruan et al. [RUA93] presented a FE head model that was partially based on geometric data of Shugar's model [SHU77]. The head model had the scalp, skull and brain as well as the dura mater and the falx cerebri incorporated. The brain was modelled as a visco-elastic material in this study. The head model was used to study direct head impact, where the simulation set-up was based on the cadaver head impact tests of Nahum [NAH77]. The intracranial pressures showed good agreement although the test conditions were not identical to those described by Nahum. The head was not rotated forward with respect to the Frankfort plane like was done during the experiment [NAH77] (annex B), resulting in a different impact location.

Besides 'validation' of the model also simulations were performed in which impactor mass and velocity and the impact position were varied. From their results the Head Injury Criterion (HIC) was found to be generally proportional with impact force and head response parameters. In a similar study Ruan et al. also varied the, linear elastic, material properties of skull, brain and cerebrospinal fluid (CSF) [RUA94]. The effect of these variations was reflected in intracranial pressure levels and skull stress and strain levels.

The same head model was used for a parametric study, in which a modal analysis was performed on the complete head model [RUA96]. The results showed that variation of skull thickness and skull element type mainly influences the two lowest natural skull frequencies. Variation of the elastic material parameters for the CSF showed no significant influence on the frequency response of the head. Inclusion of a FE neck model however, resulted in a significant decrease of the lower natural frequencies of the head. Furthermore, the inclusion of the dura and falx had a significant influence on the frequency response of the brain alone, and the dural membrane showed to a stiffening effect on the brain.

In 1997 an another head impact study was presented [RUA97], in which again Nahum's frontal impact experiment [NAH77] was simulated. Like in the earlier studies [RUA93, RUA94] agreement was found between experimental and numerical results. Now local head response parameters such as skull vonMises stress, intracranial pressure and brain principal shear strain were evaluated in relation to different types of head injury (skull fracture, focal brain injury, DAI). Variation in these results, and in HIC values, as a result of variation of impact severity was found realistic. It was concluded that their head model is suitable as a tool in head injury assessment. Furthermore HIC was again found to be a reasonable injury severity index in this study, which seems logical since in Nahum's experiment the head accelerations are mainly translational. Presence of the falx and tentorium was found to have an influence on intracranial pressure distributions. The authors concluded that for the evaluation of subdural hematomae (SDH) bridging veins and other mayor vessels inside the head need to be included in the head model.
2.3.2.2 Zhou’s FE head model (WSU brain injury model)

In 1995, Zhou et al. [ZHO95] presented a detailed FE model that is based on an earlier developed 2D porcine head model and the human head model developed by Ruan. In comparison with Ruan’s model, extra features such as grey and white matter, the ventricles inside brain and the bridging veins were incorporated in this model. Furthermore the brain mesh was refined. Like Ruan’s head model this head model was used for simulation of the Nahum’s cadaver experiment [NAH77]. All head structures were modelled as linear elastic materials in this study. Because of the agreement found between experimental and numerical intracranial pressure data (+10% overestimation of maximum contre-coup pressure), the head model was considered partially validated.

Besides to Nahum’s impact load, the model was also subjected to an angular acceleration pulse in the sagittal plane. Incorporation of an inhomogeneous brain (grey and white matter) and inclusion of the ventricles resulted in significant differences in shear stress distributions in the brain. From the simulation results it was concluded that coup/contre-coup injuries might be pressure induced while injuries in the corpus callosum and brainstem might be caused by shear strain. Shear levels in the corpus callosum (genu) were believed to be a predictor of Diffuse Axonal Injury (DAI). The response of the bridging veins was found to be in line with findings of SDH in experimental tests on rhesus monkeys. Evaluation of tensile strains in the bridging veins gave indications that the potential risk of rupture was particularly high during the brain’s rebound phase after frontal impact. Impact direction was thus expected to have an important role in the occurrence of SDH.

In 1996 a study was presented in which the same head model was used for investigation of the visco-elastic brain response to sagittal and lateral rotational head motion [ZHO96]. The visco-elastic material parameters were deducted from experimental data of Shuck and Advani [SHU72]. A parametric study was performed in which the decay factor and the white matter shear moduli were varied. Also comparison was made with elastic brain response. The results showed significant influence of decay factor variation (70 vs 700 sec\(^{-1}\)) on the shear stresses in the genu, but the influence on the shear strains was much lower. A 25% increase of the white matter short term shear modulus showed only minor influences on both shear stresses and strains. Furthermore the conclusion was drawn that lateral head rotation results in higher shear stresses in the genu of the corpus callosum and higher bridging vein strains, than sagittal head rotation with similar severity.

2.3.3 Université de Strasbourg

2.3.3.1 First FE head model

Willinger et al. presented a finite element head model in 1995 [WIL95]. This model consists of the skull, brain, falx and tentorium and the sub-arachnoid space, filled with CSF. The CSF was modelled as a solid elastic layer. NMR data were used to generate a realistic geometry of the head. The head model was used for a modal analysis in which the Young’s modulus of the sub-arachnoid space layer (CSF) was varied. This was done in order to fit the numerical modal response to the experimentally determined first natural frequency [WIL90], being approximately 100 Hz. In this same study the results of this numerical modal analysis of the head were used for constructing a prototype dummy head model.
In 1996 the same FE head model was subject to an evaluation study [WIL96]. In this study five cadaver tests were simulated. During these simulations the skull was modelled as a rigid structure. The head was loaded via prescribed skull velocity curves and intracranial accelerations and pressures were compared with experimental data. In all numerical response curves oscillations were found that were not present in the experimental results. Besides this the acceleration response showed a reasonable agreement with the experimental data. The pressure response however, showed more differences with the experiments. Only the trends in the pressure curves were found similar.

Variation of the elastic material properties of the sub-arachnoid layer showed a significant influence on the oscillations in the response curves. It was concluded that the material properties may have been chosen inaccurately and that the modelling assumption made when the sub-arachnoid layer was modelled might also be invalid. The authors found a change to visco-elastic brain material properties not to have much influence on the numerical response, which is logical since their visco-elastic time constant (28.6 ms) was nearly twice as large than the simulated time interval (15 ms). The authors conclude that the head model in its current state is not able to predict brain responses and that specific attention should be paid to the skull-brain interface in the future.

2.3.3.2 ULP head model

In 1997 an improved head model was presented [KAN97], which is named the ULP head model. In comparison with their previous head model, this model had a scalp included and the skull was modelled with three element layers, resembling the inner and outer table and the diploë. The skull was modelled as an elastic brittle material, which enabled skull fracture to be evaluated. The brain was modelled as a visco-elastic material and the parameter values were adapted from Shuck et al. [SHU72]. The material parameters for the sub-arachnoid layer were identical to those used in their previous head model. In spite of their recommendation the skull-brain interface was not really modified.

Also the ULP head model was evaluated by simulating Nahum’s cadaver head impact test [NAH77]. The peak head acceleration following the impact was found to be approximately 20% lower than the experimental peak level. The pressure response from the head impact simulation however, showed good agreement with the experimental pressure data. From the results it was concluded that the ULP head model was suitable for being used for assessing brain injury mechanisms in motorcycle accidents.

A motorcycle accident was reconstructed and head loading during the accident was simulated using the ULP head model. In order to investigate the influence of modelling a rigid skull instead of a brittle elastic skull, Nahum’s experiment was simulated with both skull versions. Since the results from this comparison were not significantly different, a rigid skull was used in the accident simulation. In this simulation the head was loaded via prescribed skull velocity curves. The authors related brain contusion found in the victim’s brain with vonMises stresses in the FE model brain. The presentation of the results does not clearly show however, whether the vonMises stress levels found at the injury region are also higher than in other (non-injured) regions of the brain.
2.3.4 Technische Universität Berlin

In 1995 Krabbel and Appel presented a CT/NMR-data based FE skull model that has a very detailed and realistic geometry and thus also quite a refined mesh. In 1996 this model was upgraded to complete head model by adding a homogeneous visco-elastic brain, two brain-surrounding layers and an interface that de-couples the brain from the skull [KRA96]. A typical feature incorporated in this model is the prescribed pressurisation on the brain surface elements in order to model the intracranial pressure of a seated human.

Later the FE head model was evaluated [KRA98] via simulation of low severity, head impact tests on human cadavers. Since no intracranial parameters were measured during these experiments, the numerical results have not been compared with quantitative experimental data, but only qualitatively linked to skull fractures and brain injuries found after the experiments and to their possible mechanisms. The calculated skull stresses, intracranial pressures and vonMises stresses were evaluated in relation with the occurrence of skull fractures and brain injuries in the cadaver experiments. The model was able to show that padding of the impactor reduced peak level skull deformations but did not reduce the average brain deformation level. This was in agreement with the fact that in both the padded and the unpadded impacts brain injuries were sometimes found.

The head model was also used to simulate a head impact against an A-pillar with and without padding. The results indicated that padding of the A-pillar reduced skull deformation to a level below fracture risk. This time also deformation of the brain was reduced significantly, which indicated a better performance of the A-pillar padding in comparison with the impactor padding.

2.3.5 National Highway Traffic Safety Administration

This FE head model was developed in 1991 with the objective of estimating strains in the brain tissue in response to dynamic forces and/or acceleration loads on the head [DIM91]. The head model has a simplified geometry and only the skull, the cerebrum and the dura mater with the falx cerebri have been included. An interface, allowing relative motion and separation, was included between the dura mater and the cerebrum. The model was used for simulating padded and unpadded A-pillar impacts. Resultant accelerations were compared with experimental data and the responsive principal and shear strains in the visco-elastic cerebrum were evaluated.

In 1994 the model was used for studying the cerebral response to translational and rotational head accelerations [BAN94]. For this study the head model was given a rigid skull and the dura-brain interface was replaced with a tie-break contact algorithm. Also an adapted, Kelvin-based, visco-elastic brain material was incorporated. The simulations were used to evaluate a newly introduced injury
criterion, named 'cumulative damage strain measure' (CSDM), which is believed to indicate the risk for diffuse axonal injury (DAI) in the brain. This CSDM measure, which is based on the maximum principal stress, turned out to be especially sensitive to rotational accelerations and much less to translational accelerations. This is in contrast with HIC, which is only sensitive to translational acceleration of the head.

In 1995 the NHTSA model was used in a study where the head loading during two different crash tests was simulated [DIM95]. Translational and rotational head velocities derived from the experiments were prescribed to the rigid skull in these simulations. Simulations with only the rotational velocities prescribed showed higher cerebral strain levels (and higher CSDM measure values) than simulations with only the translational velocities prescribed. Since HIC is not sensitive to rotational accelerations it was concluded that the CSDM measure was able to indicate the risk for types of brain injuries (DAI) that could not be predicted by HIC. Also in 1995 an anatomically detailed skull model derived from CT imaging was presented [BAN95]. Although intracranial contents were added to the model as well, this model was only used for skull fracture assessment and not for brain injury assessment.

2.3.6 Eindhoven University of Technology

At Eindhoven University of Technology a finite element head model was developed that consists of a skull, the brain, and the facial bones [CLA97a]. The geometry of this model is based on CT images of a human male skull. The head model was evaluated via modal analysis and simulation of Nahum's cadaver head impact test [NAH77]. The results showed similarities between experimental and numerical responses, although pressure peak levels were overestimated (especially the contre-coup pressure). The cause of the differences between Nahum's experimental data and the numerical response could not be addressed to one specific limitation of the model.

In 1997 this model was used to perform a parametric study in order to determine what parameters are of significant influence on the brain response [CLA97b]. For this parametric study several adapted versions of the head model were developed. Explicit modelling of intracranial substructures (cerebrum, cerebellum, brainstem, falx cerebri and tentorium cerebelli) did not induce significant changes in the brain response. Allowing relative motion between skull and brain lead to much larger changes in this response. Furthermore, variation of the Young's modulus of the brain showed to have a significant influence on the brain response.

2.4 Discussion

2.4.1 Material modelling

The head models developed since 1990 vary in geometric and anatomic detail from rather simplified models having the intracranial contents modelled as one homogeneous structure, to very complex models even having small details in the skull geometry and specific bridging veins incorporated. Although the anatomically very detailed and complex head models look more realistic, they do not
necessarily produce more realistic results when simulating loading of the human head. The biofidelity of constitutive models and interface definitions also play a critical role.

Most head models have isotropic, linear elastic or visco-elastic material models incorporated for all head structures. As long as skull fracture is not relevant in the head load simulations a linear elastic model for the skull seems sufficiently realistic. In the models that include the CSF, it is modelled as a linear elastic solid material. This is done because the Lagrangian finite element codes that are used, are not suited for modelling fluid flow. Clearly this results in non-realistic behaviour of the CSF in the head models. Also meningeal structures in the head models have linear elastic material models. Although in reality their mechanical behaviour might be more complex, in case of short duration (impact) loading a linear elastic material model is considered sufficiently accurate.

Experimental research on brain tissue has indicated that apart from visco-elastic effects, the brain behaves as a physical non-linear material (stiffness varying as a function of strain) [EST70, ARB95, MIL97]. The fact that linear (visco-)elastic material parameters used for modelling brain tissue in the different models also show a large spread (annex A), also indicates that these material models may be too much simplified for describing brain tissue behaviour.

Also the near incompressibility of brain tissue makes the use of linear (visco-)elastic material models less suitable. Nearly incompressible material modelling results in a decrease of accuracy of the simulation results, especially in combination with a relatively coarse mesh [BAT96]. In order to improve the accuracy of the brain response, in the foramen region and in general, one should either significantly refine the brain mesh, or use mixed formulation elements (both displacements and pressures as degrees of freedom).

The first option, a significant refining of the brain mesh, would result in considerably higher CPU times for the simulations. It is clear that further refining of the brain meshes in the current head models would cause a sharp increase of CPU costs, which will probably make them unsuitable for performing larger numbers of head load simulations. The second option, using mixed formulation elements for the brain mesh, would be the best manner for proper modelling of the near incompressible brain tissue [BAT96]. However, in most commercial explicit FE codes this type of element formulation is not included. This mixed formulation method will also lead to an increase in CPU costs for the head load simulations to be performed, since the number of independent variables in the model increases.

The problem of decreased accuracy of the brain potentially exists for all FE head models in which the brain is modelled with a limited amount of displacement-based elements and a linear (visco-)elastic material model. All the models reviewed in this literature study can be put this category. More sophisticated brain material models that are not affected by this potential problem, have not yet been applied successfully in head impact studies.

2.4.2 Modelling of the interfaces between the head structures

The interfaces between the different head structures are in most cases permanent tyings, including the skull-brain interface. The NHTSA head model and the TUB head model are the only models discussed here which relative motion allowed between dural and brain surface. In these models free interfaces allow both sliding motion and separation (gap forming) [DIM91, KRA98]. The tie-break interface in the NHTSA model [BAN94] includes a resistance against gap forming (which itself is more realistic), but it has the disadvantage that non-physical failure stress parameters need to be specified. In other models, such as the WSU models [RUA97, ZH095] and the ULP head model [KAN97] the CSF is included between skull and brain by means of a soft elastic layer of solid elements. Although this does allow limited relative skull-brain motion, actual fluid flow is not modelled and any significant relative motion will automatically lead to severe deformation of the CSF elements resulting in questionable numerical accuracy.
In the most realistic case one would model the CSF layer with a fluid constitutive model and in this way allow brain motion relative to the skull and also obtain realistic damping of skull-brain interactions. However, in order to model fluid flow as well as solid deformation in one model, a mixed Eulerian-Lagrangian finite element code is needed. Currently no head models have yet been developed in such a code. Clearly, in all current head models the skull-brain interface is strongly simplified and the interface conditions are described inaccurately.

2.4.3 Validation of the head models

Although many of the head models are claimed to be either fully or partly validated, in our opinion none of them is validated sufficiently for having real confidence in the biofidelity of their responses. For a thorough evaluation of the brain response not only subdural pressures (as in the Nahum simulations), but also brain deformation patterns should be proven in accordance with experimental findings. Furthermore the simulated brain response should be proven biofidelic repeatedly for different types of head loading. However, until now the acquisition of such data has proven to be very cumbersome or even nearly impossible. Currently, the experiments performed by Nahum et al. in 1977 [NAH77], are still one of the very few available sources of intracranial response data. Consequently, this source is being used for head model evaluation by most research groups. Only Willinger et al. used other intracranial response data for their head model evaluation [WIL96].

Besides head impact data also modal characteristics of the human head have been used to evaluate the modal response of some of the existing FE models of the brain, skull and complete human head. Experimental modal analyses of in vivo heads, cadaver heads and dry human skulls have been performed and published by several research groups, also in the recent past [TZE93, WIL90]. However, until now head model evaluation based on modal analysis data was limited to the comparison of natural frequencies, the eigenmodes were not evaluated.

Clearly, the main problem that remains with the evaluation of 3D FE head models is the lack of suitable experimental data. The intracranial impact response data that is available, is limited to pressures and accelerations of only few locations inside the skull and also the number of experiments published is very limited. Obtaining more experimental data that is suitable for head model evaluation should have a high priority in head injury research.
3. Configurations of the TUE and NHTSA head models

3.1 Introduction

In the framework of ADRIA the head loads suffered by five victims during frontal car crashes will be simulated. For these simulations we have three finite element head models available, being two versions of the TUE model [CLA97A, CLA97B] and the NHTSA model [DIM91, BAN94, DIM95]. In chapter four, simulations with these models will be presented and discussed, and one of the head models will be selected as the model to be used for the simulation of the victim head loads. In this chapter we will present the configurations of the head models in more detail and we will briefly discuss them.

In section 3.2 the two versions of TUE model will be presented, being a global version and a more detailed version. The global version was used as reference model in the head impact study performed by Claessens [CLA97A, CLA97B]. The detailed version is an adapted head model that was used in order to investigate the influence of the incorporation of falx and tentorium in the head model. The NHTSA head model will presented in section 3.3. In section 3.4 some features of the two models will be compared and discussed.

3.2 The TUE head models

Figure 3-1: The two versions of the TUE head model. Left: a 3D view on the global model. Right: a 3D view on the detailed model. (Parts of skull and brain have been omitted.)
3.2.1 General

The TUE head models were developed in the implicit FE code MARC [MAR94] using the preprocessor MENTAT. The global version of the TUE model includes the skull, a facial bone and the intracranial contents in the form of a homogeneous brain. The detailed TUE model has the intracranial contents divided into different substructures, being the cerebrum, falx cerebri, tentorium cerebelli, cerebellum and brainstem. Furthermore the foramen magnum is incorporated, through which the brainstem exits the skull. Both TUE models do not have the scalp incorporated. The two versions of the TUE head model are shown in figure 3-1. An overview of configurations of the two TUE models is given in table 3-1.

<table>
<thead>
<tr>
<th>Human head structures</th>
<th>Global TUE model inc. # elm.</th>
<th>elm./interface type</th>
<th>Detailed TUE model inc. # elm.</th>
<th>elm./interface type</th>
</tr>
</thead>
<tbody>
<tr>
<td>scalp</td>
<td>-</td>
<td></td>
<td>-</td>
<td></td>
</tr>
<tr>
<td>cranium (skull)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>viscerocranium (facial bones)</td>
<td>+ 156 solid elements</td>
<td></td>
<td>+ 188 solid elements</td>
<td></td>
</tr>
<tr>
<td>neurocranium</td>
<td>+ 600 solid elements</td>
<td></td>
<td>+ 2834 solid elements</td>
<td></td>
</tr>
<tr>
<td>- foramen magnum</td>
<td>-</td>
<td></td>
<td>+ opening in skull base</td>
<td></td>
</tr>
<tr>
<td>meningeal layers &amp; CSF</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>dura mater</td>
<td>-</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>- falx cerebri</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>- falx cerebelli</td>
<td>-</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>- tentorium cerebelli</td>
<td>-</td>
<td></td>
<td>+ 240 solid elements</td>
<td></td>
</tr>
<tr>
<td>arachnoid mater</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>pia mater</td>
<td>-</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>cerebro-spinal fluid (CSF)</td>
<td>-</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>brains</td>
<td>+ 1000 solid elements</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>cerebrum</td>
<td>(included in homogeneous brain)</td>
<td>+ 7568 solid elements</td>
<td>(included in homogeneous brain)</td>
<td>+ 7568 solid elements</td>
</tr>
<tr>
<td>corpus callosum</td>
<td>(included in homogeneous brain)</td>
<td></td>
<td>(included in cerebrum)</td>
<td></td>
</tr>
<tr>
<td>cerebellum</td>
<td>(included in homogeneous brain)</td>
<td>+ 730 solid elements</td>
<td>(included in homogeneous brain)</td>
<td>+ 730 solid elements</td>
</tr>
<tr>
<td>brainstem</td>
<td>(included in homogeneous brain)</td>
<td>+ 252 solid elements</td>
<td>(included in homogeneous brain)</td>
<td>+ 252 solid elements</td>
</tr>
</tbody>
</table>

Table 3-1: Configurations of the global and detailed TUE models: presentation of the structures included, the number and type of elements used for these structures, and the interface conditions between them.

3.2.2 Modelling of the skull

The skull has been modelled in the TUE models by two separate structures: the neurocranium and the viscerocranium (the facial bones). The neurocranium consists of one layer of solid elements and its geometry was determined via CT scans. Local thickness variations are incorporated in the neurocranium, although due to the limited mesh fineness small geometric details are not included. Frontal and sphenoid sinuses are not incorporated in the model. The foramen magnum is only incorporated in the detailed version of the model.

The facial bone is modelled very globally in both TUE models and its main function in the model is making the inertial properties of the model more realistic. For this reason the facial bone was given a different mass density than the neurocranium. Like the neurocranium, the facial bone consists of one layer of solid elements. Different bone layers as the inner and outer table and diploë in between are thus not explicitly modelled, making the skull a homogeneous structure. Claessens [CLA97B] modelled the skull as an isotropic linear elastic material. The material parameters were averaged from those used in other FE models in literature (see annex A).
3.2.3 Modelling of the skull-brain interface

The interface between the skull and the brain was modelled as a tied connection allowing no relative displacement between outer brain surface and inner skull surface. No intermediate layers, such as the dura mater or the CSF filled subdural/sub-arachnoid space, are incorporated in the TUE head models. Although the dura mater itself is not modelled, the falx cerebri and the tentorium cerebelli (being folds of the dura mater separating the two cerebral hemispheres and the cerebrum and cerebellum respectively) are incorporated in the detailed TUE head model. Both these structures are meshed with solid elements. The falx consists of two element layers being separated by the mid-sagittal plane and the tentorium consists of one layer of elements. Claessens [CLA97B] modelled these structures as linear elastic materials (see annex A), with the material parameters chosen equal to those used by Ruan et al [RUA91].

3.2.4 Modelling of the brain

In the global TUE model the brain is modelled as one homogenous structure, which is directly tied to the skull in a continuous mesh. The detailed model was developed starting from a refined mesh of the global TUE model. On the basis of anatomical atlas data, falx and tentorium were incorporated in the brain mesh, which automatically divided the cerebral hemispheres and separated the cerebellum from the cerebrum. Cerebral, tentorium and cerebella elements were reassigned to form the brainstem. The foramen magnum was included in the skull by reassigning some skull elements to the brainstem. All brain structures in the detailed TUE model consist of solid elements and are connected to each other, the meningeal structures and the skull in a continuous mesh. The manual way of meshing has resulted in rather simplified geometry’s and also in the introduction of degenerated 6 node solid elements, particularly near the tentorium.

Fissures, sinuses and/or ventricles within and between the brain structures are not incorporated in the TUE models. Also no distinction between white and grey matter was made. Claessens modelled the brain structures as isotropic, linear elastic materials in the TUE models [CLA97B] (see annex A). In the detailed TUE model the cerebrum and cerebellum had identical material descriptions. The brainstem was modelled more compressible, with a Poisson’s ratio of 0.40 instead of 0.48 for the other brain structures.

3.3 The NHTSA head model

3.3.1 General

The NHTSA head model was developed in the explicit Finite Element code LS-Dyna3D [LSD96]. It is a geometrically simplified model that consists of the skull, the cerebrum and an interface layer forming the dura, including the falx cerebri. The NHTSA model mesh is shown in figure 3-2. The element mesh consists completely of solid elements. In table 3-2 the configuration of the NHTSA head model is presented.

3.3.2 Modelling of the skull

In the NHTSA head model the skull is modelled as one structure including the cranial vault and the cranial base. The geometry of the facial bones is not realistically modelled in this structure (see figure 3-2). The cranial vault is modelled with three layers of solid elements. Its geometry is smooth and very much simplified. All realistic geometric details in the skull have been left out and local thickness variations have not been incorporated in the mesh. With a thickness varying between 5 and 20 mm the

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3 This is the case for the model version presented in the 1994 article [BAN94], in the original version the cranial vault was modelled as one layer of solid elements.
cranial vault becomes gradually thicker towards the cranial base. The skull mesh below the brain is completely solid, resulting in a non-realistic skull base thickness and geometry. The skull bone was originally modelled as a homogeneous, isotropic linear elastic material [DIM91] (see annex A). In later applications the skull was modelled as a rigid body [BAN94, DIM95].

Figure 3-2: The NHTSA FE head model. Left: a 3D view of the complete model. Right: a 3D view of the cerebrum.

<table>
<thead>
<tr>
<th>Human head structures</th>
<th>NHTSA head model</th>
<th>N elements</th>
<th>elm./interface type</th>
</tr>
</thead>
<tbody>
<tr>
<td>scalp</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>cranial (skull)</td>
<td>+</td>
<td>2288</td>
<td>solid elements</td>
</tr>
<tr>
<td>viscerocranium (facial bones)</td>
<td>-</td>
<td>(included in skull)</td>
<td></td>
</tr>
<tr>
<td>neurocranium</td>
<td>-</td>
<td>(included in skull)</td>
<td></td>
</tr>
<tr>
<td>- foramen magnum</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>meningeal layers &amp; CSF</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>dura mater</td>
<td>+</td>
<td>1180</td>
<td>solid elements</td>
</tr>
<tr>
<td>- falx cerebri</td>
<td>-</td>
<td>(formed by extension of the dura)</td>
<td></td>
</tr>
<tr>
<td>- tentorium cerebelli</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>- arachnoid mater</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>pia mater</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>cerebro-spinal fluid (CSF)</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>brain</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>cerebrum</td>
<td>+</td>
<td>2404</td>
<td>solid elements</td>
</tr>
<tr>
<td>corpus callosum</td>
<td>-</td>
<td>(included in cerebrum)</td>
<td></td>
</tr>
<tr>
<td>cerebellum</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>brainstem</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>interface conditions between the head structures</td>
<td>skull-dura interface: tied</td>
<td>dura-cerebrum interface: free [DIM91] / tie-break [DIM95]</td>
<td></td>
</tr>
</tbody>
</table>

Table 3-2: Configuration of NHTSA head model: presentation of the structures included, the number and type of elements used for these structures, and the interface conditions between them. (free interface = sliding motion and separation allowed.)

3.3.3 Modelling of the meninges and the skull-brain interface

The dura mater is incorporated in the NHTSA model, as a single layer of solid elements between the skull and the cerebrum. The falx cerebri, dividing the cerebral hemispheres, is also included as a part of the dura mater. It is modelled with four element layers and with 5 mm it is much thicker than in reality. The reason for this was to avoid numerical problems due to sharp elements at the tip of a very thin falx [BAN94]. The tentorium cerebelli is not included. The dural tissue, including the falx, was modelled as a linear elastic material by DiMasi et al. and Bandak et al. [DIM91, BAN94, DIM95] (see annex A). Because of the oversized thickness the stiffness, originally derived from data in literature, was reduced by a factor ten. The outer dura surface has been tied to the inner surface of the skull by means of a tied
interface prohibiting sliding motion and separation. The inner dural surface was originally separated from the outer cerebral surface by a free interface, allowing both sliding motion and separation [DIM91]. Later, this free interface was replaced by a tie-break interface [BAN94]. This interface type initially forms a tied connection, which locally releases as soon as a failure criterion is met. This criterion is described with the following inequality:

\[
\left[ \frac{|\sigma_n|}{\sigma_n^*} \right]^2 + \left[ \frac{|\tau|}{\tau^*} \right]^2 \geq 1
\]  

(3-1)

in which \(\sigma_n\) is the interface tensile stress, \(\tau\) is the interface shear stress, \(\sigma_n^*\) is the critical tensile stress for failure and \(\tau^*\) is the critical shear stress for failure. As soon as the tying has failed, it is replaced by a frictionless sliding interface with normal traction as a function of gap distance. Specific values for the critical stresses and the traction versus gap distance function are not presented in literature. A central ring of elements under the cerebrum has been connected to the with a tied interface, representing the boundary conditions imposed in reality by the brainstem exiting the skull through the foramen magnum.

3.3.4 Modelling of the brain

The cerebrum is the only brain structure incorporated in the NHTSA head model. Cerebellum and brainstem are not included; the bottom surface of the cerebrum borders on the skull base through the dura mater. The kinematic boundary conditions imposed by the brainstem exiting the skull via the foramen magnum were modelled by a local tied interface between dura and cerebrum. The interface conditions between cerebrum and dura mater are discussed in the previous section. The cerebrum is meshed completely by solid elements and is modelled as a homogeneous structure; no ventricles and cerebral substructures are included. The constitutive model that was originally used for the cerebrum was a linear visco-elastic model (Flügge model) [DIM91] (see annex A). Later a Kelvin based visco-elastic model was used [BAN94].

3.4 Discussion

3.4.1 Geometric detail and FE mesh

The TUE models and the NHTSA model show a clear difference in the level of geometric detail. The skull and brain geometry of the NHTSA model is much more simplified compared to that of the TUE models. A disadvantage of this simplified geometry is that locations inside the NHTSA model can not be correlated accurately with locations inside a real human head. On the other hand, the more realistic geometry in the TUE models also results in a more irregular element mesh of the head structures. This is seen in the brain mesh near the sphenoid sinus and in the region where the facial bone is connected to the neurocranium. In the detailed TUE model irregular meshing is seen due to the incorporation of the intracranial substructures. Especially elements in the tentorium region have become more deformed and besides 8 node solids, also degenerated 6 node solid elements and tetrahedral elements had to be included here. Irregularities in the head model mesh and inclusion of degenerated elements is also seen in most other geometrically detailed head models [KAN97, ZHO95]. Although the influence of these irregular and degenerated elements can not easily be quantified, it is clear that their influence on the accuracy of the brain response is negative.

3.4.2 The skull

In both the TUE models and the NHTSA model, frontal and sphenoid sinususes have not been incorporated. Claessens investigated the influence of including the frontal sinus in the TUE skull, and found it to be insignificant [CLA97B]. He did not investigate the influence of having the sphenoid sinus modelled. Like the earlier version of the NHTSA skull [DIM91], the TUE skulls consist of only one layer of solid elements. With a one layered skull it is possible to model skull bending correctly, only when full (8 point) element integration is used, as was done by Claessens. When reduced element
integration is used the skull should be modelled with more than one layer of solid elements, or with shell elements. However, a disadvantage of using shell elements is that the skull must be of a uniform thickness in that case.

The foramen magnum is only explicitly modelled in the detailed version of the TUE model. In general the foramen magnum is considered important in modelling brain response to head impact. It is believed to function as a pressure release mechanism when CSF flows through the foramen magnum in or out of the cranium during mechanical head loading. Also the brain, being connected to the spinal cord, is subject to specific kinematic and dynamic boundary conditions in the foramen magnum region. In the NHTSA model a tied contact definition was used at the foramen magnum region in order to incorporate more realistic kinematic boundary conditions. In the detailed TUE model, the influence of the foramen magnum is limited due to the tied skull-brain interface and the lack of modelling CSF flow. For really being able to model the pressure release mechanism the flow of CSF through the foramen magnum should be modelled.

Linear elastic modelling of the skull is quite general in FE head models. In most head impact studies a linear elastic constitutive model for the skull is considered sufficiently realistic. The main disadvantage of the linear elastic skull material models however, is that skull fracture can not be simulated. Only the risk to skull fracture can still be evaluated, by comparing bending stresses in the skull with experimentally determined skull fracture stress tolerances.

3.4.3 The skull-brain interface

The skull-brain interface is modelled in more detail in the NHTSA model than in the TUE models. The TUE models suffice with a direct and tied connection between skull and brain, where the NHTSA head model has the dura mater incorporated in combination with a contact algorithm between the dura mater and the cerebrum. Different from most other recent FE head models is that in the NHTSA model the dura mater and falx have been modelled with solid elements. Also in the detailed TUE model the falx and tentorium are modelled with solid elements. In other models the meninges are often modelled with 2D membrane/shell elements, where a solid element layer represents the (CSF filled) subdural or subarachnoid space [KAN97, RUA93, ZHO95]. The use of 2D elements seems more logical for modelling membrane structures. However, modelling the CSF in a Lagrangian code with a solid material model is also not a realistic representation of reality. Allowing relative motion between dura mater and brain by including a contact algorithm, seems a more realistic representation of reality. However, the problem remains what type of contact algorithm to incorporate and what parameter values for friction, damping and resistance against separation to use. Also the tie-break interface in the NHTSA model has not proven to be realistic and leaves parameters for the failure criterion to include for which accurate values are not really known.

3.4.4 The brain

The visco-elastic material model that characterises the NHTSA brain tissue is more realistic than the elastic model used in the TUE models (see annex A). Not only the viscous effects that are incorporated, but also the material parameters are in better agreement with reality. Especially the much higher elastic bulk modulus (1.86 GPa) used in the NHTSA model seems more realistic. Lin et al. [LIN97] determined the longitudinal wave velocities in brain tissue, from which he derived bulk moduli even higher than 2 GPa (2.41 GPa for white matter and 2.28 GPa for grey matter). With 8.3 MPa in the TUE brain the bulk modulus is underestimated with more than a factor 250. The high bulk modulus in the NHTSA brain, in combination with shear moduli that are ver much lower (34.5 kPa for short term and 17.2 kPa for long term effects in the NHTSA model) represent better the nearly incompressibility of brain tissue. The problem that remains is that linear elastic or visco-elastic material models are not very suitable for modelling nearly incompressible materials. Since better suitable material models for characterising brain tissue are not available in the FE codes that are used for head impact studies, linear elastic and visco-elastic material models are still being used at this moment.
4. Evaluation of the TUE and NHTSA head models

4.1 Introduction

Both TUE head models and the NHTSA head model have been used for head impact simulations. The first simulation set-up was derived from the set-up of one of the cadaver experiments that were performed by Nahum et al. [NAH77]. The set-up of Nahum's cadaver experiment is presented in annex B of this report. In this study validation of the head models is not the objective, since we believe more experimental data is needed for this purpose (see section 2.4.3). A comparison between experimental and numerical results will be made however, in order to get to know whether the intracranial pressures predicted by the simulations are realistic. Besides this, the main purpose of these simulations is to be able to compare the numerical responses from the different head models in case of a realistic loading of the head.

Besides simulation of Nahum's experiment, the head models will also be used for head rotation simulations. Earlier head injury studies with the NHTSA model have indicated that rotational loading of the head may lead to increased brain injury risk [BAN94, DIM95]. In the Nahum experiment, the head was not subject to severe rotational loading. The results from both types of head load simulations will be used to generally evaluate and compare the performance of the models and to investigate some model variations.

In order to enable a good comparison of the models, some features were set identical in both the TUE models and the NHTSA model. The model configurations as they were used in the simulations are presented in section 4.2. In section 4.3 the set-up and the results of the simulations of Nahum's frontal impact are presented. Section 4.4 deals with the rotational head load simulations. The results of both simulations are discussed in section 4.5. In section 4.6 this will lead to the selection of the head model most suitable for use in the victim head load simulations in the ADRIA project.

4.2 Head model configurations used for the simulations

Since most head models have been developed for explicit FE codes, and working with explicit time integration is much less expensive qua CPU costs, both TUE head model versions have been translated from the implicit MARC code into the explicit LS-DYNA3D code for this study. The NHTSA head model was already available in LS-DYNA3D code. For this study the NHTSA head model has been translated and scaled from American units into SI-units. In the following subsections the configurations of the head models are presented.

4.2.1 Meshing and element formulation

In all simulations the reduced element integration was used for all solid elements, except for the skull elements in the cases that the skull was modelled as a non-rigid body. Hourglass control is used to suppress the zero energy deformation modes that exist for reduced integration brick elements, according to the recommendations in the LS-DYNA3D manuals [LSD96].
4.2.2 Anthropometry

The dimensions of the head models have been adapted to the anthropometric measures of the cadaver head in Nahum’s head impact experiment (see annex B). This was done in HyperMesh [HYP97] by scaling height, width and length of the head models.

4.2.3 Material modelling

The material models and parameters for the different head structures during the evaluation simulations are presented in table 4-1. With the NHTSA model only rigid skull simulations will be performed. Furthermore, in the NHTSA model the meningeal tissue has a Young’s modulus that is ten times lower than that in the TUE model, which is because of the overestimated thickness of the dura in the NHTSA model (section 3.3.3). The material models for the skull, facial bone and meningeal tissue in the TUE models, are the same as those used by Claessens et al. [CLA97A, CLA97B]. The parameters for the linear visco-elastic brain model are adapted from Bandak et al. [BAN94], since they were found most realistic (section 3.3.4).

<table>
<thead>
<tr>
<th>head structure</th>
<th>constitutive model</th>
<th>ρ (kg/m³)</th>
<th>E (MPa)</th>
<th>K (MPa)</th>
<th>G (MPa)</th>
<th>τ (ms)</th>
<th>ν (-)</th>
</tr>
</thead>
<tbody>
<tr>
<td>skull</td>
<td>lin. elastic (TUE only)</td>
<td>2070</td>
<td>6500</td>
<td>-</td>
<td>-</td>
<td>0.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>rigid body</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td></td>
<td></td>
</tr>
<tr>
<td>facial bone (TUE models only)</td>
<td>lin. elastic rigid (incl. in skull)</td>
<td>5000</td>
<td>6500</td>
<td>-</td>
<td>0.2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>meningeal tissue (dura, falx, tentorium)</td>
<td>lin. elastic (TUE)</td>
<td>1130</td>
<td>31.5</td>
<td>3.15</td>
<td>0.45</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>lin. elastic (NHTSA)</td>
<td>1130</td>
<td>31.5</td>
<td>3.15</td>
<td>0.45</td>
<td></td>
<td></td>
</tr>
<tr>
<td>brain tissue (cerebrum, cerebellum, brainstem)</td>
<td>lin. visco-elastic</td>
<td>1040</td>
<td>1860</td>
<td>G₀ = 0.0344</td>
<td>Gₐ₀ = 0.0172</td>
<td>10</td>
<td></td>
</tr>
</tbody>
</table>

Table 4-1: The material models used in the TUE and NHTSA head models during our evaluation simulations. (with ρ being the mass density, E the Young’s modulus, K the bulk modulus, G the shear modulus (G₀ short term and Gₐ₀ long term), τ the visco-elastic time constant and ν the Poisson ratio).

4.2.4 Interface conditions

In the TUE models all head structures are solidly tied to each other. In the NHTSA head model the skull and the dura mater are also tied. Between the dura mater and the cerebrum the tie-break interface that was implemented by Bandak et al. is kept [BAN94]. The critical stresses for the failure criterion (equation 3-1) and the constant gap-traction force were left identical to the values that were already present in the NHTSA model (see table 4-2). After failure of the interface frictionless sliding is modelled. The locally tied interface between cerebrum and dura mater in the original NHTSA model configuration (representing the foramen magnum) is also maintained.

| Critical tensile stress (σ₀* in equation 3-3) | 101.35 kPa |
| Critical shear stress (τ* in equation 3-3)    | 34.4 kPa   |
| Gap resisting tensile stress (after failure)  | 101.35 kPa |

Table 4-2: Tie-break interface parameters for the dura-brain interface in the NHTSA head model.

4.3 Simulation of Nahum’s cadaver experiment

4.3.1 Simulation set-up

The NHTSA model and the global and detailed TUE model have been used for simulation of the experiment indicated by Nahum et al. as experiment 37 [NAH77] (annex B). In these experiments, the frontal skull was hit with a rigid impactor. Instead of explicitly modelling the impactor hitting the forehead, the impact force measured during Nahum’s experiment was used as input load data for
simulations with both TUE models. This force has been transformed into a pressure load by dividing it by the area covered by the frontal skull elements on which the pressure load is applied. The temporal variation of the impact force is approximated by a goniometric function. In figure 4-1 the simulation configuration, the impact force versus time and the derived pressure versus time are shown. The impact load was applied under a 45° angle relative to the Frankfort plane, as was done in the experiments themselves.

Apart from the input load no dynamic or kinematic constraints have been imposed on the head. This also implies that restriction of head motion by the neck is not modelled. In the short time interval that is simulated (10 ms) this influence is not expected to be very significant. The simulation set-up that is described here, is applied to the global and the detailed TUE head model with skulls being elastic deformable.

From the simulation with the global TUE model the velocities of two skull nodes (in frontal and in occipital region) were translated to velocity pulses for the centre of gravity (cg) of the skull. These cg-velocity pulses (x- and z-direction and rotation around y-axis) were used as prescribed velocity input for simulations with the NHTSA model and the global TUE model with the skulls being modelled as rigid bodies. The simulation with the rigid skull TUE model was performed in order to evaluate influence of skull deformations on the brain response. This simulation also relates better to the simulation performed with the NHTSA model, making comparison of the two models easier. Table 4-3 gives an overview of the four simulations that have been performed. Also the CPU time needed for the simulations is given. The simulations were performed on a Silicon Graphics R10000 workstation.

<table>
<thead>
<tr>
<th>FE Model</th>
<th>Skull-brain interface</th>
<th>Skull material modelling</th>
<th>Applied head loading conditions</th>
<th>Simulated time period</th>
<th>CPU costs</th>
</tr>
</thead>
<tbody>
<tr>
<td>TUE global</td>
<td>brain tied to skull</td>
<td>elastic</td>
<td>pressure load on frontal skull</td>
<td>10 ms</td>
<td>0 h. 13 min.</td>
</tr>
<tr>
<td>TUE global</td>
<td>brain tied to skull</td>
<td>rigid</td>
<td>prescribed skull cg velocities</td>
<td>10 ms</td>
<td>0 h. 02 min.</td>
</tr>
<tr>
<td>TUE detailed</td>
<td>brain tied to skull</td>
<td>elastic</td>
<td>pressure load on frontal skull</td>
<td>10 ms</td>
<td>2 h. 28 min.</td>
</tr>
<tr>
<td>NHTSA</td>
<td>dura mater + tie-break interface</td>
<td>rigid</td>
<td>prescribed skull cg velocities</td>
<td>10 ms</td>
<td>0 h. 30 min.</td>
</tr>
</tbody>
</table>

Table 4-3: Overview matrix of the Nahum simulations that have been performed with the head models. The CPU times needed for the simulations are also given.

4.3.2 Simulation results

Figure 4-2 shows the elastic skull cg-velocity curves derived from the elastic skull simulation with the global TUE model. These cg-velocity curves were used as input for rigid skull simulations.
Figure 4-2: The velocity curves that were determined from the global TUE elastic skull response and prescribed as input for the rigid skull simulations. Left: linear velocities in x- and y-direction. Right: angular velocity around y-axis.

Figure 4-3 shows the resultant accelerations of two skull nodes, located at the frontal bone (impact location, left figure) and the occipital bone (contre-coup region, right figure). The skull acceleration curve from Nahum's experiment itself is also included in both plots. Comparison of the accelerations for these two nodes enables us to check whether using the approximated elastic skull cg-velocities on the rigid skull models leads to the same loading conditions for both simulation set-ups. As can be seen the acceleration curves show quite good agreement. Only the frontal skull accelerations show some differences at their maximum levels. Comparison of the simulation results with Nahum’s experimentally determined head acceleration shows that the experimental conditions are correctly reproduced up to 3.8 ms, after which the deviations between experiment and simulations become larger.

Figure 4-3: Skull accelerations from the simulations and from the Nahum experiment. Left: acceleration of a node at impact location (frontal bone). Right: acceleration of a node at contre-coup location (occipital bone). (tue_gl = global TUE model, tue_det = detailed TUE model, nhtsa = NHTSA model, esk = elastic skull, rsk = rigid skull)

Like in Nahum’s experiment, also intracranial pressures were part of the output. In figure 4-4 pressures in the coup region (frontal) and contre-coup region (occipital) have been compared with the experimental results. The pressures were averaged for previously identified brain elements in both regions. For each of these elements the pressure was determined in the following manner:

\[ p = -1/3 \cdot (\sigma_1 + \sigma_2 + \sigma_3) = -1/3 \cdot (\sigma_{xx} + \sigma_{yy} + \sigma_{zz}) \]  

(4-1)
In this equation $\sigma$ represents the normal stresses in principal directions 1, 2, and 3 or in global direction $x$, $y$ and $z$. Next for each region the element pressures were averaged. The elements in the two regions were chosen so that in each model the pressures were determined at locations as identical as possible.

![Graph](image1.png)

**Figure 4-4:** Intracranial pressures from the simulations and from Nahum's experiment. Left: coup pressures (frontal region). Right: contre-coup pressures (occipital region). (tue_g1 = global TUE model, tue_det = detailed TUE model, nhtsa = NHTSA model, esk = elastic skull, rsk = rigid skull)

![Graph](image2.png)

**Figure 4-5:** Pressure contour plots at $t=4$ms. Sagittal cross-section through brain, 10 mm left of mid-sagittal plane. Upper left: global TUE model with elastic skull. Upper right: global TUE model with rigid skull. Lower left: detailed TUE model with elastic skull. Lower right: NHTSA model with rigid skull.

The results show more spread in the response of the different head models than was seen for the accelerations. Both the experimental coup and contre-coup pressures are to some extent overestimated by the numerical pressures. The NHTSA model peak level pressures come closest to the experimental peak levels (coup: $-\approx$equal, contre-coup: $-60\%$), whereas the detailed TUE model shows the largest overestimation of peak level pressures (coup: $-50\%$, contre-coup: $-130\%$). The response from the
NHTSA model also shows the most different response with an oscillation seen during the last four milliseconds of the simulation.

The pressure distribution inside the brain is also presented via contour plots in figure 4-5. These plots show cross-sections through the brain 10 mm left of the mid-sagittal plane. The 10 mm shift was done to prevent the cross-section from running through the falx in the detailed TUE model and the NHTSA model. The plots all show the pressure distribution at $t = 4$ ms, which is shortly after maximum head acceleration occurred. In figure 4-4 as well as in annex C (fig. C-1) it can be seen that also the pressures have just passed their maximum at this point in time. All models predict a similar, simple stress pattern with maximum pressures near the impact location (coup side) and minimum pressures in the contrecoup region. The detailed TUE model shows some differences in the pattern around the tentorium, compared with the global TUE model with elastic skull. Also a small shift in the pressure contours is seen. Modelling the skull as a rigid body in the global TUE model also resulted in a shift in the pressures, but now in the other direction (comparison of upper plots).

Figure 4-6 shows vonMises stress contour plots from the four simulations. vonMises stresses are defined as a function of the three principal stresses by equation 4-2:

$$\sigma_{\text{von}} = \frac{1}{2} \cdot \sqrt{\frac{1}{2} \cdot \left[(\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_3 - \sigma_1)^2\right]}$$

The vonMises stresses are solely related to brain tissue deformation, not to volumetric change of the brain tissue. The vonMises contour plots show the stress distributions at $t = 5$ ms. At this point the vonMises stresses are close to their maximum, as can be seen in annex C, figure C-2. Identical sagittal cross-sections as for the pressure distributions are displayed.

![Figure 4-6: vonMises contour plots at t=5ms. Sagittal cross-section through brain, 10 mm left of mid-sagittal plane. Upper left: global TUE model with elastic skull. Upper right: global TUE model with rigid skull. Lower left: detailed TUE model with elastic skull. Lower right: NHTSA model with rigid skull (different scaling of fringe levels here!)](image-url)
The upper figures show that changing the skull from elastic to rigid hardly had any influence on the vonMises stress patterns. The detailed TUE model shows more differences in the vonMises stress pattern, particularly near the brainstem and sphenoid sinus. Also in the centre of the brain (in and around the corpus callosum) the stresses have increased in comparison with the response from global TUE models. The NHTSA model shows a quite different stress pattern, not comparable with the patterns seen in the TUE models. Also the stress levels are higher here, therefore the fringe level scaling for the NHTSA model stress distribution (lower right plot) is taken twice as large as for the stress distributions in the TUE models.

4.4 Rotational loading of the head

4.4.1 Simulation set-up

Since in the Nahum simulations the head models were not subject to severe rotational loads, also rotational head loading has been simulated. In all these simulations the skull was modelled as a rigid body, so that the loading could be applied to the head models by means of prescribed rotational velocities (fig. 4-7). The rotation is in all cases applied in posterior-anterior direction, meaning rotation around the y-axis in positive direction. The time velocity curve is a sinusoidal function that results in a smoothly increasing velocity reaching its maximum at 30 ms, when the head is rotated 45° forward. Next the velocity decreases to zero ending in a final head rotation of 90° after 60 ms.

The configurations of the TUE and NHTSA head models in the rotational load simulations were identical to those used for the Nahum simulations (section 3.5). Also the anthropometric head measures were kept the same (annex B, fig. B-2). In table 4-4 an overview is given of the rotational load simulations that were performed. Besides simulations with both TUE models and the NHTSA model, an extra simulation was performed with the NHTSA model, having a free interface between dura and cerebrum instead of the tie-break interface. This free interface allowed sliding motion and separation without resistance. Comparison of the two NHTSA simulations can give us some information on the influence of skull-brain interface modelling on the brain response.

<table>
<thead>
<tr>
<th>FE Model</th>
<th>Skull-brain interface</th>
<th>Skull material modelling</th>
<th>Applied head loading conditions</th>
<th>Simulated time period</th>
</tr>
</thead>
<tbody>
<tr>
<td>TUE global</td>
<td>brain tied to skull</td>
<td>rigid</td>
<td>angular velocity around y-axis</td>
<td>70 ms</td>
</tr>
<tr>
<td>TUE detailed</td>
<td>brain tied to skull</td>
<td>rigid</td>
<td>angular velocity around y-axis</td>
<td>70 ms</td>
</tr>
<tr>
<td>NHTSA</td>
<td>dura mater + tie-break interface</td>
<td>rigid</td>
<td>angular velocity around y-axis</td>
<td>70 ms</td>
</tr>
<tr>
<td>NHTSA</td>
<td>dura mater + free interface</td>
<td>rigid</td>
<td>angular velocity around y-axis</td>
<td>70 ms</td>
</tr>
</tbody>
</table>

*Table 4-4: Overview matrix of the rotational load simulations that have been performed with the head models.*
4.4.2 Simulation results

The output from the rotational head load simulations can not be compared with experimentally determined accelerations and pressures, as was done for the Nahum simulations. Therefore only vonMises stress contour plots will be presented here. The vonMises stresses are more interesting than pressures in these simulations, due to the rotational and non-impact nature of the head loading. In figure 4-8 vonMises stress contour plots are presented for each simulation at $t = 20$ ms. At this moment the (inertial) brain loading has just passed its maximum (fig. 4-7, right plot). In annex C (figure C-2) it can be seen that also the vonMises stresses are close to their maximum at this moment.

![Image of vonMises stress contour plots](image)

Figure 4-8: vonMises contour plots at $t=20$ms. Sagittal cross-section through brain, 10 mm left of mid-sagittal plane. Upper left: global TUE model. Upper right: detailed TUE model. Lower left: NHTSA model with tie-break interface. Lower right: NHTSA model with free interface.

The vonMises stress plots show that the two TUE models show a roughly similar stress pattern, with high stresses near the superior sagittal sinus and the sphenoid sinus. Their responses differ most significantly near the tentorium cerebelli and in the brainstem. Also the response from the NHTSA model with the tie-break interface shows similarities with that from the global TUE model. Finally the free cerebrum-dura interface in NHTSA model resulted in a significantly different brain response than the response seen tie-break interface NHTSA model. Both the pattern has changed and the stress levels have become higher.

4.5 Discussion and evaluation

4.5.1 Numerical versus experimental response

Only the results of the Nahum simulations could be compared with experimental data. Nahum’s head acceleration pulses are approximated quite reasonably in the first four milliseconds. After that the decelerations in the simulations start later than in the experiment. The nearly constant acceleration seen
next in the experiment is not predicted in the simulations. The nature of this constant acceleration in the experiment is not known; possibly it represents influences of the neck.

The pressures pulses show that the coup-contrecoup effect (positive pressures in the coup region and negative pressures in contrecoup region) seen in the experimental results, is also predicted by the simulations. In both regions the pressure levels are overestimated though, particularly in the occipital region. This is probably because the flow of CSF around the brain (which has a damping effect on the brain motion) is modelled within the head models. Comparison of the experimental and numerical pressure responses after four milliseconds is not useful, since the accelerations are not accurately simulated after this point in time.

Although more accurate prediction of the experimental results would be desirable, it should be kept in mind that we are comparing our results with the limited data from only one experiment, giving us no information on the reliability of these data. The simulations do show us that the nature of the experimental response can be predicted with the head models.

4.5.2 Rigid versus elastic skull modelling

The effect of modelling the skull as a rigid body instead of an elastic deformable body could be evaluated from the two Nahum simulations with the global TUE model. As explained in section 4.3 the kinematic output of elastic skull simulation was used to derive the prescribed velocity input for the rigid skull simulation (fig. 4-2). The only real difference between the simulations is that the elastic skull also undergoes deformation following the impact force on the forehead. This skull deformation can be identified in the frontal skull node accelerations (fig. 4-3), where near the peak acceleration the elastic skull behaviour shows some deviation from the rigid skull behaviour.

The elastic skull deformation also explains the higher pressures seen in comparison with the rigid skull model: due to this deformation the total intracranial volume is reduced slightly, leading to extra pressure on the brain. This overall pressure increase for the elastic skull simulation can also be seen in the pressure-time graphs in figure 4-4. Since the pattern of the stress contours did not change and the vonMises stresses were hardly affected by making the skull rigid, it can be concluded that modelling the skull either elastic or rigid does not significantly influence the brain response patterns to frontal head impact.

In both cases focal brain injuries directly resulting from local skull fracture can not be simulated. The only advantage of the elastic skull is that via calculated local skull bending stresses the risk to fracture could be estimated, which is not possible with a rigid skull.

4.5.3 Influence of modelling intracranial substructures and foramen magnum

Evaluation of the influence of including intracranial substructures such as falx cerebri and tentorium cerebelli and the foramen magnum, was done by comparison of the results of the global and detailed TUE model simulations. Both the Nahum and the rotational loading simulations could be used. Although the mesh being coarser in the global TUE model than that in the detailed model might also cause some differences in the brain responses, these effects are not expected to result in significant changes in the stress patterns than are seen.

The influence of the falx cerebri is not very obvious. In the pressure responses from the Nahum simulations (fig. 4-5) no clear changes are seen in the falx region. The higher vonMises stresses (fig. 4-6) in the upper region of the brain in the detailed model might be caused by the presence of the falx, but also the refined mesh in itself could have some influence here. The increase is not seen in the rotational loading simulations. The influence of the falx cerebri is expected become more obvious in case of head rotations out of the sagittal plane.
The influence of the tentorium cerebelli is more clearly seen. Particularly in the rotational loading case the presence of the tentorium in the detailed model obviously has a reducing effect on the vonMises stresses above and underneath the structure (fig. 4-8). Furthermore the tip of the tentorium, where the splenium of the corpus callosum is situated (fig. 4-6 and 4-8) caused a local increase of vonMises stresses. Also the pressure distributions in the Nahum simulation show influence of the presence of the tentorium, although this influence is very little.

The effect of modelling of foramen magnum is quite obvious in the vonMises stress patterns of both simulation set-ups. In both cases an increase of vonMises stresses is seen in the brainstem. This increase is the largest in the Nahum simulations. Whether the effects of the inclusion of the intracranial substructures are realistic can only be evaluated with a validation procedure (see section 2.4.3).

In the detailed TUE model also a petrous bone structure is included on the skull base. In this model the tentorium is connected to this petrous bone. Right anterior of this petrous bone a gap in the mesh is seen in the detailed TUE brain mesh (fig. 4-5, 4-6, 4-8). Both the existence of this gap and the geometry of the petrous bone structure are not realistic and they also seem to affect the vonMises stresses in this region (fig. 4-6, 4-8).

4.5.4 Influence of variations in skull-brain interface modelling

For evaluation of the influence the skull-brain interface conditions have on the brain response, the NHTSA model simulations are the most useful. In the rotational load case two versions of the NHTSA model were used: one with the tie-break interface three and one with a free interface. When we compare the vonMises stress contours plots for these simulations (fig. 4-8) we see large differences. The free interface results in a more complex stress pattern and, even more obvious, much higher vonMises stress levels overall (maximum vonMises stress more than three times higher). In the tie-break interface almost no failure of the contact was seen, which means that the interface remained tied almost everywhere in the interface. The tie-break NHTSA model shows, in spite of the differences in brain geometry, quite reasonable agreement with the global TUE model as far as the pattern is concerned (fig. 4-8). This agrees with the fact that both models have more or less a tied interface.

In the Nahum simulations, the tie-break interface in the NHTSA model showed much more failure of the initial tying. For this reason the differences between the vonMises stress patterns in the global TUE model and the NHTSA model are much larger. In the regions where failure occurred an increase of the vonMises stresses is seen. The influence of failure is also seen in the occipital pressure pulses in figure 4-4, where the point of failure in this region is clearly visible (t ~ 2.4ms). Like in the free interface model in the rotational loading case the vonMises stresses are higher overall (note the difference in fringe level scaling in figure 4-6).

In general it can be seen from the simulations that the interface conditions between brain and dura mater have a significant effect on the brain response. Since it was already concluded that no realistic models of the skull-brain interface have yet been developed in any head model (see section 2.4.2), this is an important conclusion. It indicates that implementation of more realistic skull-brain interface models in FE head models should have a high priority in numerical head injury research.

4.6 Conclusions and recommendations

Conclusions following from the simulation results:

From the evaluation simulations performed here the following conclusions could be drawn:

- Reasonable agreement found between experimental and numerical pressure responses indicates that the nature of the pressure response of the head models in this study is realistic. The exact
pressure levels should not be considered as accurate predictions of reality. The biofidelity of the vonMises stress response of the brain could not be evaluated with experimental data.

- Modelling the skull as a rigid body instead of an elastic deformable structure does not significantly influence the brain response patterns in terms of pressure and vonMises stresses; only a slight overall decrease of pressures is seen.

- Inclusion of falx cerebri and tentorium cerebelli does have its influence on the brain response. Therefore they should be included in any realistic FE model of the human head.

- The way of modelling the skull-brain interface conditions has proven to have a very significant influence on the brain response. However, the simulation results do not indicate what type of interface conditions result in the most realistic brain response.

**Recommendations concerning set-up of the victim head loading simulations in the ADRIA project:**

The fact that making the skull rigid does not significantly affect the brain response is an important conclusion with respect to the set-up for the victim head load simulations to be performed in the ADRIA project. It allows us to use prescribed skull velocity pulses (in six degrees of freedom) as input for the simulations instead of applied impact forces to the skull. Application of impact force loads would significantly complicate the simulations. In that case boundary effects from head-neck interaction would have to be modelled in order to certify a correct representation of the loading conditions.

In the head load simulations to follow in the ADRIA project, the input velocity pulses are to be obtained from the head centre of gravity kinematics, which form part of the output of the MADYMO3D accident reconstruction simulations.

**Recommendations concerning the FE head model to be used for the victim head loading simulations:**

The results from the Nahum simulation with the different head models have not been able to prove that one of the head models provides significantly more biofidelic brain responses than the others. For this reason we will have to select the head model on other criteria, such as geometry and anatomic detail.

From the simulations with the head models we were able to conclude that inclusion of a falx cerebri and a tentorium cerebelli in the head model did have an influence on the brain response. This makes the detailed head model more suitable than the other two head models. Further advantages of the detailed TUE model are a more realistic geometry than the NHTSA head model and a more refined mesh than the global TUE model.

A disadvantage of the detailed TUE model is the CPU costs, which are higher for this model than for the others. In spite of this we have decided to use the detailed TUE model for the victim head load simulations to be performed in the ADRIA project. However, before these simulations can be performed, some features of the model need to be improved:

- In the sphenoid sinus the mesh needs to be adapted in order to remove the existing gap in the cerebral mesh and make both the skull (petrous bone) and the cerebral geometry more realistic.

- In regions where highly deformed elements are present, the mesh should be made more regular. The presence of degenerated (6 node) solid elements should be avoided where possible.
- The skull-brain interface could be made more realistic. Inclusion of the dura mater as an
intermediate layer between skull and brain enables implementation of other skull-brain interface
conditions, like included in the NHTSA head model.

- Other interface conditions between dura mater and brain (sliding motion with or without the
possibility of separation) can then be incorporated in the TUE model as model variants to the tied
interface. Possible increase or decrease in correlation between numerical response and medical
data concerning the victim’s brain injuries could then indicate what type of interface conditions
result in the most realistic brain response.
5. Improvements to the detailed TUE head model

5.1 Introduction

In the previous chapter it was decided to use the detailed version of the TUE head model for simulation of the head loads suffered by the car crash victims studied in the ADRIA project. However, before performing these simulations some improvements have been made to the head model. The improvements concern the quality of the element mesh of the model as well as the anatomy and geometry of the model in certain regions. After these improvements were made to the model, the influence of the skull brain interface conditions in the TUE model on the brain response has been studied. For this study a sliding interface between skull and brain was implemented in the new head model. The improvements made to the model and the evaluation of them are described in respectively sections 5.2 and 5.3. Section 5.4 deals with the skull-brain interface study.

5.2 Mesh and geometry modifications

During our evaluation of the TUE head model it was found that the geometry and the mesh needed some improvement, particularly the region of the sphenoid sinus. In the detailed TUE model the petrous bone inside the skull was originally modelled as a wall structure emerging from the skull base (see figure 5-1, left).

![Figure 5-1: Mid-sagittal cross-sections of the skull: view on the skull base of the original TUE model (left) and the modified TUE model (right).](image)

Implementation of this petrous bone resulted in an unrealistic geometry of the temporal cerebral lobes, anterior to this ‘wall’. The anatomical atlas shows that in reality the sphenoid sinus, which is not included in the original TUE model, is situated in this region [MIN94]. Based on this anatomic atlas, both the skull and cerebrum in this region have been remodelled in order to make their geometry more realistic (see figure 5-1, right). During the remodelling process also the sphenoid sinus has been included in the head model.

Besides in the sphenoid sinus region also in the region just superior to the tentorium the cerebrum has been remeshed. In the detailed TUE model the tentorium itself crosses diagonally through the cerebral
elements, resulting in many degenerated (penta- or tetrahedral) elements. Also the tentorium geometry
did not have a realistic smooth surface, as can be seen in figure 5-2 (left). In the modified TUE model
this surface has been smoothed. Furthermore the cerebral region superior to the tentorium has been
remeshed in order to remove all the degenerated solid elements. The result of this can also be seen in the
falx mesh in figure 5-2 (right). Geometrical changes of the falx cerebri were made particularly the
superior and the frontal region. Furthermore, the falx cerebelli, a dural fold into the cerebellum, is
included in the modified TUE model.

![Figure 5-2: View on the falx and the tentorium of the original (left) and the modified TUE model (right). The dura mater, which is included in the modified TUE model, is not shown in the right figure.](image)

Because the falx and tentorium are in fact folds of the dura mater, the complete dura mater is also
implemented in the modified TUE model. In the modified model the outer edges of the tentorium and the
falx, seen in figure 5-2 (right), connect to the rest of dura mater, which is not shown in the figure. In the
original model this was not the case: here the falx cerebri and tentorium cerebelli were modelled as two
separate structures inside the neurocranium, being connected directly to the inner skull surface.

### 5.3 Evaluation of the mesh and geometry modifications

The modifications made to the TUE model have been evaluated afterwards. The evaluation consisted of
a check of the FE-mesh quality, a check on the inertial properties of the model and the performance of
head loading simulations with the modified head model.

#### 5.3.1 Quality of the FE-mesh

After having changed the intracranial mesh, of course we want to know whether the FE mesh of the
modified head model has actually improved in comparison with the mesh of the original TUE head
model. For this purpose the 8 node solid elements in both meshes have been checked on their maximum
warpage angles, their aspect ratio’s, their maximum skew angles and their Jacobian values. For the
manner in which these entities are determined one is referred to the HyperMesh User’s manual
[HYP97].

The skull elements have not been included in the check, since the skull will be modelled as a rigid body
in the head load reconstruction simulations in the ADRIA project. In figure 5-3 the results of the
element checks of both models are compared in histograms. The results show that for all four criteria
the FE mesh of the modified head model has improved in relation to that of the original TUE model.
Normally an improved mesh will also lead to numerically more accurate simulation results.
5.3.2 Inertial properties

Head length (195 mm), width (155 mm) and height (225 mm) of the model are scaled to 50\(^{th}\) percentile measures according to Pheasant [PHE86]. With these measures and the mass density of the (schematically modelled) facial bone chosen 4.5 times as large as the skull mass density (see table 5-1), we obtain a total head mass of 3.87 kg. The difference between the mass of our model and the 50\(^{th}\) percentile human head mass of approximately 4.55 kg can be explained by the fact that in our model the scalp is not modelled. When looking at for instance the WSU brain injury model [ZHO95] we see that in this model the scalp mass is 0.66 kg. When we add this mass to the total mass of our head model we obtain a new total mass of 4.53 kg, which is in much better agreement with the 4.55 kg for the 50\(^{th}\) percentile human head.

The head model has been defined in the global co-ordinate system with its anatomical origin located in the global origin. The x- and y-axis span the Frankfort plane with the x-axis pointing in anterior (forward) direction and the z-axis in superior (upward) direction. The head’s anatomical origin (a.o.) was defined relative to the position of the occipital condyles (o.c.), for which a corresponding node of the mesh was chosen on the basis of drawings of the human body [TRI80]. These drawings also define the locations of the anatomic origin and the centre of gravity of the 50\(^{th}\) percentile human head. It turned out that with the mass densities chosen as in table 5-1, the calculated location of the centre of gravity differed less than 1 cm from the theoretical location on the human body drawings. The good approximations of both the head mass and centre of gravity indicate that the corrected mass density for the facial bone was accurately chosen in combination with these particular mass densities for the other head structures.

In the ADRIA project the head load simulations will be performed via prescribed motions on a rigid skull. In this case the inertial properties of the skull will not have any influence on the brain responses.
Only the inertial properties of the intracranial contents are then relevant. When we look in the modified FE head model we find that the total volume of the intracranial contents equals 1463 ml. With 1.36 kg the total brain mass in the model is also sufficiently realistic.

<table>
<thead>
<tr>
<th>head structure</th>
<th>number of elements</th>
<th>mass density (kg/m³)</th>
<th>mass (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>skull</td>
<td>3212</td>
<td>2.333</td>
<td></td>
</tr>
<tr>
<td>neurocranium</td>
<td>3024</td>
<td>1.477</td>
<td></td>
</tr>
<tr>
<td>facial bone</td>
<td>188</td>
<td>0.856</td>
<td></td>
</tr>
<tr>
<td>meningeal tissue</td>
<td>3188</td>
<td>0.172</td>
<td></td>
</tr>
<tr>
<td>dura mater</td>
<td>2536</td>
<td>0.133</td>
<td></td>
</tr>
<tr>
<td>falx cerebri</td>
<td>448</td>
<td>0.022</td>
<td></td>
</tr>
<tr>
<td>falx cerebelli</td>
<td>18</td>
<td>0.001</td>
<td></td>
</tr>
<tr>
<td>tentorium cerebelli</td>
<td>186</td>
<td>0.016</td>
<td></td>
</tr>
<tr>
<td>brain tissue</td>
<td>7692</td>
<td>1.363</td>
<td></td>
</tr>
<tr>
<td>cerebrum</td>
<td>6758</td>
<td>1.126</td>
<td></td>
</tr>
<tr>
<td>cerebellum</td>
<td>732</td>
<td>0.201</td>
<td></td>
</tr>
<tr>
<td>brainstem</td>
<td>202</td>
<td>0.036</td>
<td></td>
</tr>
<tr>
<td>total</td>
<td>14092</td>
<td>3.868</td>
<td></td>
</tr>
</tbody>
</table>

Table 5-1: Review on the new TUE head model (masses valid for 50th percentile head measures).

5.3.3 Simulation of the Nahum experiment

In order to evaluate the performance of the modified TUE model, the Nahum experiment has been simulated with both the original detailed TUE model and the modified TUE model. The simulation setup is identical to the one used in chapter 4; the prescribed velocities that are shown in figure 4-2 are applied to the rigid skulls of both head models. Also the material characteristics have been chosen identical as in chapter 4 and can be found in table 4-1.

Figure 5-4: Brain response from the Nahum simulation: pressure time-history curves. Frontal pressures are shown left and occipital pressures are shown right. Plotted are the experimental results from Nahum et al, and the simulation results from the original and the modified TUE head model.
Results:

Figure 5-4 shows the pressure time histories for the modified TUE model, the original detailed TUE model and Nahum's experiment. The curves show that the modified model predicts somewhat lower peak pressure levels in both the frontal and the occipital region. Furthermore it can be seen that the original model shows a positive occipital pressure during the last three milliseconds of the simulation, where in the modified model this positive pressure is not present. In both simulations the frontal as well as the occipital pressure amplitudes are overestimated.

Figure 5-5 shows the pressures (upper plots) and vonMises stresses (lower plots) in the plane 10 mm left of the mid-sagittal plane in the original (left) and modified TUE model (right). In the sagittal cross-sections the modifications made to the tentorium and the sphenoid sinus region are clearly illustrated. The pressure distributions in the two models show a shift in the pressures indicating an overall pressure reduction in the modified model. Apart from this the stress patterns are quite similar. Also the global vonMises stress patterns have not changed very much in the modified model. The vonMises stress distributions in both cases show local maxima near the foramen magnum and sphenoid sinus. In the modified model the vonMises stresses decreased near the sphenoid sinus and increased near the splenium of the corpus callosum (just anterior to the tip of the tentorium).

Discussion:

The reduction of the frontal pressure level in the modified model can be explained by the inclusion of the dura mater between skull and brain, which reduces the acceleration applied by the rigid skull to the outer brain surface to some extent. The elastic behaviour of the dura mater in the model can be seen in figure 5-4, in the form of a minor oscillation superposed to the pressure curves from the modified model. In reality the flow of CSF may result in a even further reduction of the pressure amplitudes, which would explain the overestimation of the pressure amplitudes in our simulations. The difference in
the occipital pressures deserves some more study. In order to determine the occipital pressures in the TUE models, in each hemisphere four elements were used to average the pressures from. In figure 5-6 the pressures for the elements in the left hemisphere are displayed for both models. In the original model (left) two of these four elements are 6-node degenerated elements, which turn out to give quite a different pressure history than the 8-node elements. In the modified model the mesh has been changed in order to remove the degenerated elements. As can be seen in figure 5-6 (right) this resulted in a much more similar pressure curves for the four elements. For this reason the occipital pressures from the modified model seem more reliable than those from the original model. This is in spite of the fact that the curve of the original model seems more realistic during the last 3 ms of the simulations.

The most obvious change seen in the vonMises stress contour plots is the decrease of the stress levels near the sphenoid sinus. This is a direct result of the modifications made in this region of the modified model. The local maximum of vonMises stresses that is still seen in this region is probably due the presence of the falx cerebri, of which the tip is located quite nearby. For the same reason a local stress maximum is situated near the splenium of the corpus callosum, being quite near the tip of the tentorium cerebelli.

The increase of vonMises stresses near the foramen magnum is still present in the modified TUE model. These high stresses are the result of relatively large deformation of the brainstem elements within the foramen magnum. This is caused by the combination of a relatively coarse mesh in this region, the tied dura-brain interface (skull-brain interface in the original model) and the free lower end surface of the brainstem. Since the tied-dura brain interface is known not to be very realistic, with the current lower brainstem configuration the predictions in this region should be handled with care. In combination with a tied dura-brain interface, restricting the motion of the lower end surface nodes of the brainstem might be more realistic and reduce the vonMises stresses locally.
5.4 Variation of skull-brain interface conditions

5.4.1 Characterisation of the interface conditions

The modified TUE head model has been used for studying the effects of variations in the skull-brain interface conditions. In reality the skull-brain interface is a quite complex system with several meningeal layers lying between the cortex of the brain and inner skull surface. These meningeal layers have cerebrospinal fluid (CSF) flowing in between them. Furthermore there are the bridging veins crossing the different layers of the interface. For the interface conditions this means that sliding motion between the meningeal layers is possible but limited due to the bridging veins, where they cross these layers.

In described in section 5.2 in the modified head model the dura mater, which is situated in between the skull and the brain, is incorporated. The interface between the dura mater and the outer brain surface in the model represents the subdural space, where in reality CSF is flowing and sliding motion between brain and dura mater is possible. In the head model as used so far this sliding motion is not allowed, since the brain was tied to the dura mater.

5.4.2 Modelling of the interface conditions

In order to study the effect of permitted sliding motion between brain and dura mater we have incorporated different contact prescriptions for the brain - dura mater interface than the permanent tying used so far. Two different contact prescriptions for the brain-dura interface have been studied: a free interface and a sliding interface. In table 5-2 the characteristics of these interface types are given.

<table>
<thead>
<tr>
<th>Interface type:</th>
<th>resistance against separation:</th>
<th>friction against sliding:</th>
</tr>
</thead>
<tbody>
<tr>
<td>Permanently tied (&quot;TIED&quot; in LS-Dyna3D)</td>
<td>infinite</td>
<td>infinite</td>
</tr>
<tr>
<td>Sliding (&quot;SLIDING_ONLY_PENALTY&quot; in LS-Dyna3D)</td>
<td>infinite</td>
<td>zero</td>
</tr>
<tr>
<td>Free (&quot;SLIDING&quot; in LS-Dyna3D)</td>
<td>zero</td>
<td>zero</td>
</tr>
</tbody>
</table>

Table 5-2: Brain-dura interface types and their characterisation.

In the sliding interface separation of the dura from the brain is not allowed, only sliding between the two structures is possible. The free interface allows both sliding between and separation of brain and dura mater. A tie-break version of the interface, as was used in the NHTSA head model [BAN94], is not included in this study, because we consider inclusion of the unknown, non-physical parameters for the failure criterion (equation 3-1) in the head model undesirable. The head model with tied interface option is considered the reference model in this study. The sliding and free contact algorithms both use the penalty method in the LS-Dyna3D FE code. In this study all friction and damping parameters for the contact algorithms have been set to zero.

5.4.3 Evaluation

5.4.3.1 Simulation of the Nahum experiment

The different interface conditions have been evaluated by again using the Nahum simulation set-up that was used in sections 4.3 and 5.2 of this report. The material parameters used in the head model are given in table 5-2. In figure 5-7 the frontal and occipital pressure response curves are plotted. It can be seen that the non-tied interfaces cause oscillations in the pressure response. Particularly the free interface causes very large peaks in the signals. Mainly due to the high bulk modulus of the brain the brain-dura contact is rather stiff, resulting in high penalty factors for penetrating nodes. Since no damping is applied in the contact this stiff contact will cause oscillations normal to the contact surface. In the free interface these oscillations are clearly more severe than in the sliding interface. In this study
we have not further investigated the influences of contact damping. At this moment we are primarily interested in the global effects of the interface conditions on the brain response.

The most obvious difference in the pressure response can be seen in the occipital region, where the free interface results in pressures that remain close to the zero level and do not become negative at all. Also the sliding interface reduces the negative peak pressure level (-25%), but the negative pressure pattern remains similar to the tied interface reference case. In the frontal pressures we see the free interface resulting in a significantly higher peak pressure level (averaged peak level ±70% higher).

![Figure 5-7: Brain response from the Nahum simulation: pressure time history curves. Frontal pressures are shown left and occipital pressures are shown right. Plotted are the experimental results from Nahum et al, and the simulation results from the modified TUE head model with respectively tied, sliding and free brain-dura interface. (For the oscillating signals from the non-tied interfaces also filtered curves are shown.)](image)

Also the sliding interface shows a higher peak pressure level, but here the difference with the tied interface is only 10%. The differences in the pressure curves in figure 5-7 are illustrated once more in the para-sagittal pressure contour plots in figure 5-8. The free interface shows a shift in the pressure distributions so that even in the occipital region no more negative pressures can be observed. The sliding interface results in a much less severe shift, although in this case some variation in the pressure pattern can be observed.

The vonMises stress patterns (right plots in figure 5-8) also indicate a significant influence of the interface conditions on the brain response. Both the free and the sliding interface cause an increase of vonMises stresses in the superior sagittal sinus region. Also in the regions of falx and tentorium tip the vonMises stress has become larger with respect to the tied interface case. The overall global vonMises patterns do not seem to have changed significantly, except for a significant increase of the vonMises stresses in the cerebellum in the free interface case.

**Discussion:**

The results from the Nahum simulations clearly show a significant influence of the brain-dura mater interface conditions on the brain response. The fact that a free interface results in non-negative contre-
coup pressures is easily explained, since there is no force that resists the brain from separation from the dura mater. In case of a sliding interface this resisting force is present, but the possibility of sliding motion does have a decreasing effect on the maximum occipital pressure with respect to the tied interface case. The fact that negative contre-coup pressures can not be predicted with a free interface makes this type of interface less suitable for modelling the brain-dura interface conditions, especially since contre-coup effects are considered a potential mechanism leading to focal brain injuries like contre-coup contusions. The sliding interface does enable the prediction of these contre-coup effects.

Pressures

![Pressures Diagram]

VonMises stresses

![VonMises Stresses Diagram]

**Figure 5-8**: Brain response from the Nahum simulation: pressure distributions (left plots) and vonMises stress distributions (right plots). Variation of interface conditions: tied interface (upper plots), sliding interface (middle plots) and free interface (lower plots). The results are shown in a plane 10 mm left of the mid-sagittal plane.
5.4.3.2 Rotational head load simulation

In order to further investigate the behaviour of the sliding brain-dura interface a second simulation has been performed, in which a rotational loading was applied to the head. The simulation set-up was identical to the one used in section 4.4 of this report and is described in figure 4-7. In this simulation the sliding interface caused big problems at the falx tip, where severe and unrealistic element deformations were seen. In a second run of the simulation, now with the falx tip being tied to the brain, the numerical problems could be avoided. When studying the response at the interface surface for this second simulation, it can be seen that sliding motion does occur along the flat side surface of the falx cerebri, but hardly takes place at the curved and less smooth brain-dura interface. This is illustrated in figure 5-9, where both the undeformed and the deformed state of the brain and dura mesh during head rotation have been illustrated.

Figure 5-9: Brain response from rotational head load simulation: deformed (left) and undeformed (right) brain mesh. Mid-sagittal cross-section of the brain and surrounding dura mater (falx cerebri is not shown).

Discussion:

The results shown in figure 5-9 indicate that the response of the outer brain surface is very sensitive to the positioning of nodes on the interface surface, which is in fact a simplification of the brain surface in reality. Due to this phenomenon, unrealistic shear stress distributions at the brain surface can be expected. It also disabled sliding motion along the falx tip (when it was not yet tied in the first simulation attempt), what caused the numerical problems during the simulation.

The best method to overcome this problem would be the use of quadratic elements of the brain and dura layer and in this way making the interface surface smooth. However, this would lead to a large increase of CPU time. When taking into account that the simulation described here already cost nearly 19 hours of CPU time, it is clear that implementation of quadratic elements would make the head model unsuitable for use in studies with large numbers of head load simulations. For this reason no further effort has been spent to the implementation the sliding interface in the TUE head model, and a tied brain-dura interface is maintained.
6. Summary and conclusion

6.1 Summary

In this study we have evaluated recently developed 3D Finite Element head models in general and two of them in particular: the TUE head model and the NHTSA head model. From our literature review (section 2.2) it was concluded that at this moment three main problems are encountered when studying head impact with 3D FE head models:

1. The lack of a suitable and accurate material model for describing the mechanical behaviour of brain tissue. The linear elastic and visco-elastic constitutive models that are currently applied assume physically linear material behaviour, which is questionable for brain tissue. Furthermore, these models are not very suitable for describing nearly incompressible material behaviour.

2. The impossibility of realistic and accurate modelling of interface conditions between intracranial structures in general and between skull and brain in particular, due to the complexity of the interfaces in reality (cerebro-spinal fluid (CSF) flowing in between membranes overlying the different brain structures and bridging veins crossing these fluid layers).

3. The impossibility of thorough evaluation (and possible validation) of 3D FE head models due to the lack of experimental intracranial impact response data available. Necessary brain deformation data have not yet been acquired from head impact experiments.

In the discussion concerning the configurations of the TUE and NHTSA head models (section 3.4) it was concluded that these head models are not suitable for assessing brain injuries directly evolving from skull fracture, since skull fracture itself can not be simulated. Brain material modelling in the two head models was done using linear elastic and visco-elastic constitutive models, like in most other FE head models described in literature. Evaluation of pressure wave propagation speeds showed that the brain material parameters in the in the NHTSA model were chosen more realistic than in the TUE models. As far as geometry is concerned the NHTSA model, with the cerebellum and tentorium cerebelli not included, is more simplified than the TUE model. On the other hand, the skull-brain interface in this model is modelled in more detail (inclusion of dura mater and contact algorithms) than in the TUE models (permanently tied connection).

The TUE and NHTSA head models were further evaluated via simulations of Nahum’s experiment and simulations of rotational loading of the head (sections 4.3 and 4.4). Brain material parameters in both models were set equally to those from the NHTSA model, since they were found most realistic. From the comparison of Nahum’s pressure data with the numerical responses it could be concluded that the models showed reasonably realistic responses. The nature and to a certain extent the order of magnitude of the intracranial pressure pulses was in agreement with the experimental data.
The evaluation simulations also showed that modelling the skull rigid instead of elastic had no significant effect on the brain response (section 4.5). Pressure and vonMises stress distributions did not change except for a slight global increase (offset) of the pressures due to elastic skull deformation. The influence of including falx and tentorium in the model was shown in both simulation set-ups, although the effect was best seen in the rotational load simulations. With the NHTSA model the influence of the brain-dura interface conditions was shown. Replacement of the tie-break interface with a free interface resulted in very significant changes in the vonMises stress patterns in the brain.

Based on the more realistic geometry and the presence of intracranial substructures like cerebrum, cerebellum, brainstem, falx and tentorium, the detailed TUE model was considered best suitable for use in the ADRIA project (section 4.6). A number of modifications were made to this model, including mesh and geometry adaptations in most of the incorporated head structures (section 5.2). Furthermore the dura mater was included as a solid layer in between the brain and the skull. The skull was modelled as a rigid body, so that prescribed head centre of gravity velocities, derived as output from the accident reconstructions can be applied to the skull as input load. Evaluation of the modifications (section 5.3) showed that they were an improvement to the model, both in computational accuracy and geometrical correctness.

A study on the brain-dura interface conditions in the improved TUE model (section 5.4) showed that, like in the NHTSA model, the brain response was significantly affected by replacement of the tied interface by a sliding or a free interface. The free interface proved not to be able to predict negative contre-coup pressures in the brain response. The sliding interface caused numerical problems leading to unrealistically large deformations at the falx tip. Also a large sensitivity of the interface surface mesh on the response of the outer brain surface (cortex) was induced with a sliding interface. For these reasons it was decided not to use other interface models than the tied interface in the TUE model during the victim head load simulations to be performed in ADRIA.

6.2 Conclusion

We will conclude this report with presenting the configuration of the finite element head model, as it will be used for the simulation of victim head loads in the ADRIA project. The new TUE head model is shown in figure 6-1. The model consists of a continuous mesh, built up completely with solid elements (table 5-1). The three main structures of the model are the skull, the meninges and the brain. The brain is modelled with a linear visco-elastic model and the meninges with a linear elastic model. The skull is modelled as a rigid body, on which head velocity in six degrees of freedom will be prescribed as input head loads. For this purpose the centre of gravity of the skull is set in the theoretical location of the head centre of gravity. This location, relative to the head's anatomic origin, is based on drawings of the human body [TRI80].

With the choice of applying the head loads in the form of prescribed velocities on a rigid skull, we eliminated the possibility of evaluating the response of the skull to head impact loading. However, the alternative of explicit modelling objects impacting a deformable skull would introduce other complications in the head load simulations. Accurate boundary conditions from the neck then need to be included and also the facial structures need to be modelled more realistically, particularly in the case of facial impacts (e.g. to the steering wheel). For this reason, and because the skull being rigid instead of elastic did not change the brain response significantly, it was decided to choose for rigid skull modelling.
In table 6-1 the material properties defined for the meningal and brain structures are presented. The brain material parameters are set identical to those used earlier in NHTSA model [BAN94], which were found more realistic than those earlier applied in the TUE models (section 3.4.4). The material parameters for the meningal structures are set identical those used earlier in the TUE model by Claessens et al. [CLA97B]. These parameters were also used earlier by Ruan et al. [RUA91].

![Diagram of the new TUE head model](image)

*Figure 6-1: Overview of the new TUE head model. The right half of the head model is shown, with the brain, the dura mater with falx and tentorium and the skull being separated. In the model these structures are connected to each other, resulting in a continuous finite element mesh.*

<table>
<thead>
<tr>
<th>head structures</th>
<th>mass density (kg/m³)</th>
<th>bulk modulus (N/mm²)</th>
<th>short term shear modulus G₀ (N/mm²)</th>
<th>long term shear modulus Gₘ (N/mm²)</th>
<th>Young’s modulus E (N/mm²)</th>
<th>Poisson ratio v (-)</th>
<th>decay factor β (s⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>dura mater, falx, tentorium</td>
<td>1130</td>
<td></td>
<td></td>
<td></td>
<td>31.5</td>
<td>0.45</td>
<td>-</td>
</tr>
<tr>
<td>cerebrum, cerebellum, brainstem</td>
<td>1040</td>
<td>1860</td>
<td>0.0344</td>
<td>0.0172</td>
<td></td>
<td></td>
<td>100</td>
</tr>
</tbody>
</table>

*Table 6-1: Material properties incorporated in the new TUE head model (the skull is modelled as a rigid body).*

The simulations of the head loads suffered by the victims from the reconstructed accidents in the ADRIA project, will be performed with this head model in the explicit FE code LS-Dyna3D, on a Silicon Graphics Origin Workstation with R10000 processor.

The biofidelity of the brain response predictions of the presented head model could not be evaluated thoroughly due to the lack of experimental data to compare with. For this reason it is important to interpret the results of the head load simulations with great care. The quantitative aspects of the predicted response, such as specific stress or strain levels reached, should not be considered to represent the exact levels reached in the brain in reality. Therefore it is recommended to use the brain response predictions particularly for comparison of spatial stress/strain distributions with the locations of the injuries in reality and for studying the differences and similarities in the responses from different simulations in relation to the head loading conditions and HIC.
References


Transport Research Institute, "Anthropometric specifications for the mid-sized male dummy (side view with skeleton)", human body drawings: drawing no. MM-104, project-contract no. DTHH22-80-C-07502.


Annex A

In this annex data concerning the 3D Finite Element head models, reviewed in chapter 2 of this report, is tabulated. Table A-1 shows the scalp configurations in the head models. In table A-2 the skull configurations are presented, table A-3 deals with the meningeal structures and the CSF, and in A-4 the brain configurations are given. The explanations of the symbols used in the tables are presented in table A-5. For all head models only the configurations used in their last applications are presented.

In the linear elastic visco-elastic models used for the brain tissue in many head models, stiffness against volume change is modelled with an elastic bulk modulus (K). Only the shear modulus (G(t)) is modelled visco-elastic time dependent, with the use of equation A-1:

\[ G(t) = G_\infty + (G_0 - G_\infty) e^{-(t/\tau)} \]  

\[ \text{(A-1)} \]

### Table A-1: Scalp configurations in the head models reviewed in chapter 2 of this report.

<table>
<thead>
<tr>
<th>head model [ref]</th>
<th>structure (type)</th>
<th>number of elem.</th>
<th>constitutive model</th>
<th>( \rho ) (kg/m(^3))</th>
<th>( E ) (MPa)</th>
<th>( \nu ) (-)</th>
</tr>
</thead>
<tbody>
<tr>
<td>WSU model I [RUA97]</td>
<td>scalp shell</td>
<td>1111</td>
<td>lin. elastic</td>
<td>1200</td>
<td>16.7</td>
<td>0.42</td>
</tr>
<tr>
<td>&quot;WSU brain injury model&quot; [ZHO96]</td>
<td>scalp shell</td>
<td>2728</td>
<td>lin. elastic</td>
<td>1200</td>
<td>16.7</td>
<td>0.42</td>
</tr>
<tr>
<td>ULP model I [WIL96]</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>&quot;ULP head model&quot; [KAN97]</td>
<td>scalp solid</td>
<td>-</td>
<td>lin. elastic</td>
<td>1200</td>
<td>16.7</td>
<td>0.42</td>
</tr>
<tr>
<td>TUB head model [KRA98]</td>
<td>scalp shell</td>
<td>1005</td>
<td>lin. elastic</td>
<td>1200</td>
<td>34.5</td>
<td>0.40</td>
</tr>
<tr>
<td>NHTSA head model [JIM95]</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>TUE head model (detailed version) [CLAO97]</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

### Table A-2: Configurations of the skull, facial bone and neck in the head models reviewed in chapter 2 of this report. (* value represents the plastic tangent hardening modulus, ** value represents the yield stress)

<table>
<thead>
<tr>
<th>head model [ref]</th>
<th>structure (type)</th>
<th>number of elem.</th>
<th>constitutive model</th>
<th>( \rho ) (kg/m(^3))</th>
<th>( E ) (MPa)</th>
<th>( \nu ) (-)</th>
<th>UTS (MPa)</th>
<th>UCS (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>WSU model I [RUA97]</td>
<td>skull (table)</td>
<td>2554</td>
<td>lin. elastic</td>
<td>3000</td>
<td>5468</td>
<td>0.22</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>facial bone</td>
<td>240</td>
<td>lin. elastic</td>
<td>1750</td>
<td>2728</td>
<td>0.24</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>neck bone</td>
<td>140</td>
<td>lin. elastic</td>
<td>1750</td>
<td>2728</td>
<td>0.24</td>
<td></td>
<td></td>
</tr>
<tr>
<td>&quot;WSU brain injury model&quot; [ZHO96]</td>
<td>skull solid</td>
<td>2702</td>
<td>lin. elastic</td>
<td>2100</td>
<td>4000</td>
<td>0.22</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ULP model I [WIL96]</td>
<td>skull shell</td>
<td>1082</td>
<td>rigid body</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>&quot;ULP head model&quot; [KAN97]</td>
<td>skull (table)</td>
<td>-</td>
<td>elastic brittle</td>
<td>1800</td>
<td>15000</td>
<td>0.21</td>
<td>90</td>
<td>145</td>
</tr>
<tr>
<td></td>
<td>facial bone</td>
<td>-</td>
<td>elastic brittle</td>
<td>1500</td>
<td>4500</td>
<td>0.00</td>
<td>35</td>
<td>28</td>
</tr>
<tr>
<td>TUB head model [KRA98]</td>
<td>skull solid</td>
<td>13962</td>
<td>elastic-plastic</td>
<td>2075</td>
<td>2420</td>
<td>0.19</td>
<td>366**</td>
<td>100**</td>
</tr>
<tr>
<td>NHTSA head model [BAN94]</td>
<td>skull solid</td>
<td>2288</td>
<td>rigid body</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TUE head model [CLAO97]</td>
<td>skull solid</td>
<td>2834</td>
<td>lin. elastic</td>
<td>2070</td>
<td>6500</td>
<td>0.22</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>facial bone</td>
<td>188</td>
<td>lin. elastic</td>
<td>5000</td>
<td>6500</td>
<td>0.22</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

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Table A-3: Configurations of the meningal structures, the CSF and bridging veins in the head models reviewed in chapter 2 of this report. (* contact interface included between dura mater(attached to skull) and pia mater(attached to brain), ** contact interface included between dura mater(attached to skull) and cerebrum).

<table>
<thead>
<tr>
<th>head model [ref]</th>
<th>structure</th>
<th>element type</th>
<th>number of elem.</th>
<th>constitutive model</th>
<th>ρ (kg/m³)</th>
<th>E (MPa)</th>
<th>ν (%)</th>
<th>$\lambda$ (kPa)</th>
<th>$\mu$ (kPa)</th>
<th>$\tau$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>WSU model I [RA97]</td>
<td>dura, falk, tentorium CSF, sinus, fissures</td>
<td>membrane</td>
<td>1596</td>
<td>lin. elastic</td>
<td>1133</td>
<td>31.5</td>
<td>0.45</td>
<td>1040</td>
<td>1.49</td>
<td>0.4887</td>
</tr>
<tr>
<td>“WSU brain injury model” [ZH096]</td>
<td>dura, falk, tentorium pia mater CSF, sinus, ventricles bridging veins</td>
<td>membrane</td>
<td>741</td>
<td>lin. elastic</td>
<td>1133</td>
<td>31.5</td>
<td>0.45</td>
<td>1040</td>
<td>1.49</td>
<td>0.4887</td>
</tr>
<tr>
<td>ULP model I [WL96]</td>
<td>dura, falk, tentorium subarachnoid space (CSF)</td>
<td>shell solid</td>
<td>256</td>
<td>lin. elastic</td>
<td>1140</td>
<td>31.5</td>
<td>0.45</td>
<td>1040</td>
<td>0.012</td>
<td>0.49</td>
</tr>
<tr>
<td>“ULP head model” [KAN97]</td>
<td>dura, falk, tentorium subarachnoid space (CSF)</td>
<td>shell solid</td>
<td>1316</td>
<td>lin. elastic</td>
<td>1140</td>
<td>31.5</td>
<td>0.45</td>
<td>1040</td>
<td>0.012</td>
<td>0.49</td>
</tr>
<tr>
<td>TUB head model [KRA98]*</td>
<td>dura mater pia/mater cerebrum</td>
<td>shell solid</td>
<td>3985</td>
<td>lin. elastic</td>
<td>1133</td>
<td>31.5</td>
<td>0.45</td>
<td>1040</td>
<td>0.012</td>
<td>0.49</td>
</tr>
<tr>
<td>NHTSA head model [DIM95]**</td>
<td>dura mater, falks</td>
<td>solid</td>
<td>1180</td>
<td>lin. elastic</td>
<td>1130</td>
<td>3.15</td>
<td>0.45</td>
<td>1040</td>
<td>0.012</td>
<td>0.49</td>
</tr>
<tr>
<td>TUE head model [CLA97A]</td>
<td>dura, falk, tentorium subarachnoid space (CSF)</td>
<td>shell solid</td>
<td>554</td>
<td>lin. elastic</td>
<td>1130</td>
<td>31.5</td>
<td>0.45</td>
<td>1040</td>
<td>0.012</td>
<td>0.49</td>
</tr>
</tbody>
</table>

Table A-4: Configurations of the brain structures in the head models reviewed in chapter 2 of this report.

<table>
<thead>
<tr>
<th>head model [ref]</th>
<th>structure</th>
<th>element type</th>
<th>no. of elem.</th>
<th>constitutive model</th>
<th>ρ (kg/m³)</th>
<th>E (MPa)</th>
<th>ν (%)</th>
<th>$\lambda$ (kPa)</th>
<th>$\mu$ (kPa)</th>
<th>$\tau$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>WSU model I [RA97]</td>
<td>gray matter</td>
<td>solid</td>
<td>5580</td>
<td>visco-elastic</td>
<td>1040</td>
<td>2190</td>
<td>41/7.1</td>
<td>34/6.3</td>
<td>1.43</td>
<td></td>
</tr>
<tr>
<td>“WSU brain injury model” [ZH096]</td>
<td>white matter cerebellum, brainstem</td>
<td>solid</td>
<td>3284</td>
<td>visco-elastic</td>
<td>1040</td>
<td>2190</td>
<td>41/7.1</td>
<td>34/6.3</td>
<td>1.43</td>
<td></td>
</tr>
<tr>
<td>ULP model I [WL96]</td>
<td>cerebrum</td>
<td>solid</td>
<td>1316</td>
<td>lin. elastic</td>
<td>1040</td>
<td>0.857</td>
<td>0.48</td>
<td>2.45</td>
<td>2.86</td>
<td>28.6</td>
</tr>
<tr>
<td>“ULP head model” [KAN97]</td>
<td>cerebellum</td>
<td>solid</td>
<td>741</td>
<td>lin. elastic</td>
<td>1040</td>
<td>0.675</td>
<td>0.48</td>
<td>2.45</td>
<td>2.86</td>
<td>28.6</td>
</tr>
<tr>
<td>TUB head model [KRA98]*</td>
<td>brainstem, corp, callosum</td>
<td>solid</td>
<td>224</td>
<td>visco-elastic</td>
<td>1040</td>
<td>0.675</td>
<td>0.48</td>
<td>2.45</td>
<td>2.86</td>
<td>28.6</td>
</tr>
<tr>
<td>NHTSA head model [DIM95]**</td>
<td>homogenous brain</td>
<td>solid</td>
<td>9273</td>
<td>visco-elastic</td>
<td>1040</td>
<td>1.28</td>
<td>5.28/168</td>
<td>7.8</td>
<td>10</td>
<td></td>
</tr>
<tr>
<td>TUE head model [CLA97A]</td>
<td>cerebrum</td>
<td>solid</td>
<td>9288</td>
<td>visco-elastic</td>
<td>1040</td>
<td>1.28</td>
<td>5.28/168</td>
<td>7.8</td>
<td>10</td>
<td></td>
</tr>
</tbody>
</table>

Table A-5: Explanation of the symbols used in tables A-1 to A-4 for the material parameters defined in the head models.

<table>
<thead>
<tr>
<th>symbol</th>
<th>entity</th>
<th>unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\rho$</td>
<td>mass density</td>
<td>kg/m³</td>
</tr>
<tr>
<td>E</td>
<td>Young’s modulus</td>
<td>MPa</td>
</tr>
<tr>
<td>$\nu$</td>
<td>Poisson’s ratio</td>
<td>-</td>
</tr>
<tr>
<td>UTS</td>
<td>Ultimate tensile stress</td>
<td>MPa</td>
</tr>
<tr>
<td>UCS</td>
<td>Ultimate compression stress</td>
<td>MPa</td>
</tr>
<tr>
<td>K</td>
<td>Bulk modulus</td>
<td>MPa</td>
</tr>
<tr>
<td>G0</td>
<td>Short term (dynamic) shear modulus</td>
<td>kPa</td>
</tr>
<tr>
<td>G∞</td>
<td>Long term (infinite) shear modulus</td>
<td>kPa</td>
</tr>
<tr>
<td>$\tau$</td>
<td>visco-elastic time constant (1/decay factor)</td>
<td>ms</td>
</tr>
</tbody>
</table>
Annex B

B.1 General set-up of Nahum’s cadaver experiments

Nahum et al. performed head impact experiments on human cadavers in order to investigate brain response [NAH77]. During the experiments the cadavers were seated stationary when their heads were impacted with a rigid impactor. The cadavers’ cranial vascular network and the CSF space were fluid pressurised to in vivo pressure levels beforehand. The impactor mass, impact velocity and padding material on the impactor varied for each experiment. Impact occurred on the frontal bone of the skull in the mid-sagittal plane. A schematic view of the test set-up is shown in figure 4-1. This figure shows the direction and location of impact on the head relative to the head anatomical co-ordinate system. In the tests the cadaver heads were rotated forward so that the Frankfort plane was inclined 45° to the horizontal plane. The impactor struck the heads in anterior-posterior direction.

Figure B-1: Schematic view of the set-up of Nahum’s experiment (Hcg = head centre of gravity)

During the experiments dynamic measurements of impact force and bi-directional (x and z direction) head acceleration were performed. In addition intracranial subdural pressure-time histories were recorded at the following locations:

- Underneath the frontal bone, adjacent to impact area.
- Underneath the parietal bones, immediately posterior to coronal and superior to squamosal suture.
- Underneath the occipital bone, inferior to the lambdoidal suture.
- Underneath the occipital bone, at the posterior fossa.

B.2 In- and output data of Nahum’s experiment 37

Experiment 37 is the only experiment of which the results have been presented in more detail by Nahum et al. [NAH77]. For this reason this specific experiment is the one that is always used for evaluation of FE head models in literature. Also in our model evaluation procedure this specific experiment will be simulated. Therefore some specific input data of this experiment is given here.
In figure B-2 some of the anthropometric head measures of the cadaver used in experiment 37 are presented. Figure B-3 shows the impact force that the impactor applied to the frontal skull and the resultant translational acceleration of the head's centre of gravity. The subdural pressures measured underneath the frontal bone (near impact location) and underneath the occipital bone (inferior to the lambdoidal suture) are shown in figure B-4.

Figure B-2: Anthropometric measures of the head of the cadaver that was used in experiment 7.

Figure B-3: Output data of Nahum's experiment 37. Left: Impactor force versus time. Right: Resultant linear acceleration versus time of the head centre of gravity.

Figure B-4: Output data of Nahum's experiment 37. Left: Coup pressure versus time. Right: Contrecoup pressure versus time.
C.1 Pressure and vonMises stress contour plots from the Nahum simulation with the rigid skull global TUE model.

In figure C-1 the pressure contour plots from the Nahum simulation with the global TUE model with rigid skull are displayed. The plots show a cross-section through the brain taken parallel to and 10 mm left of the mid-sagittal plane. Contour plots are shown for $t = 0.0$ ms up to 9.0 ms with a time interval of 1 ms. The plots visualise the time history of the pressure in the TUE brain. In a similar way the vonMises stress contour plots are displayed in figure C-2.

![Figure C-1: Pressure contour plots from the Nahum simulation with the rigid skull global TUE head model. Cross-sections 10 mm left of mid-sagittal plane. Plots for $t = 0.0$ ms to $t = 9.0$ ms with time interval of 1 ms (sequence left to right). (Fringe levels: black = maximum pressure level: > 150 kPa; white = minimum pressure level: < -150 kPa.)(Image not provided in this text)}
C.2 vonMises stress contour plots from the rotational load simulation with the rigid skull global TUE model.

For the rotational load simulation the vonMises stresses are visualised via contour plots as well. Figure C-4 shows these contour plots for $t = 0\text{ ms}$ to $t = 70\text{ ms}$ with a time interval of $10\text{ ms}$. Also in this case the cross-sections through the brain were taken parallel to and $10\text{ mm}$ left of the mid-sagittal plane.
Figure C-3: vonMises stress contour plots from the rotational load simulation with the rigid skull global TUE head model. Cross-sections 10 mm left of mid-sagittal plane. Plots for $t = 0$ ms to $t = 70$ ms with time interval of 10 ms (sequence up to down). (Fringe levels: black: maximum vonMises stress: > 3.0 kPa; white: minimum vonMises stress: < 0.0 kPa.)