Finite element modeling of head impact: The second decade.

A.A.H.J. Sauren and M.H.A. Claessens

Eindhoven University of Technology, Faculty of Mechanical Engineering,
P.O. Box 513, 5600 MB Eindhoven, The Netherlands.

Abstract

A review is given of papers on finite element modeling of head impact that have been published in the period 1982-1992. The main conclusions as to the progress made in this decade can be summarized as follows. The constitutive models used are predominantly linearly elastic or viscoelastic, the former almost all in combination with small-deformation theory and the latter with large deformations. Only one model included both non-linear viscoelasticity and large deformations. Utilization of this type of sophisticated models continues to be hampered by the lack of consistent and complete experimental data. Successful efforts have been made to implement more realistic boundary conditions allowing relative motion between the skull and the brain. Boundary conditions at the head-neck junction need further investigation. The same holds for the application of scaling methods not only to models with different geometries but also to similarly-shaped models with different boundary conditions at the interfaces between the substructures. More and more use is made of physical models both for validating finite element modeling techniques and as a starting point for parametric studies.

1 Introduction

In head injury research, like in many areas of biomechanics, mathematical tools from the field of applied mechanics have proved their value for obtaining a better understanding of the mechanical behavior of the complex system comprised by the cranial vault and its contents.

The mathematical models used to analyze the response of the head to impact or inertial loading can be subdivided into three types; lumped parameter models, continuum models and finite element models. The main limitation of the lumped parameter models is constituted by the fact that - inherent in their discrete nature - they do not lend themselves to describing the distribution of field parameters in the model, such as stress, strain and pressure. With continuum models closed-form solutions can be obtained describing field parameter distributions. However, these models are only tractable after introducing idealizations concerning geometry, constitutive properties and boundary conditions. Finite element models are in principle capable of handling complex geometries and different kinds of nonlinearities. In the early seventies finite element modeling entered the research field of head biomechanics. To date it is commonly recognized as a powerful and versatile tool for exploring and analyzing many different aspects of the mechanical behavior of the head under extreme loading conditions.

In an extensive and thorough review of finite element models of head impact Khalil and Viano [1982] delineated the state of the art in the early eighties. They pointed out some critical features of the models proposed up to that time, which can be briefly summarized as follows. In the majority of the models the skull-brain system was approximated by a fluid-filled shell that lacked the possibility of relative motion between shell and fluid. A proper formulation of boundary and interface conditions was hampered by the requirement of continuity of stress and displacement between elements. Another important issue was the use of homogeneous, isotropic and linearly (visco)elastic material models based on small deformations in spite of the fact that biological tissues in general are nonhomogeneous and anisotropic materials, subject to large deformations and having non-linear time-dependent constitutive properties. Finally large discrepancies were observed between the frequency responses of the shell models and data obtained from experiments with real shells.

Since Khalil and Viano reviewed the first decade of finite element modeling of head impact, another decade has elapsed. The present paper is an attempt to delineate the developments made during this second decade.

2 Overview and brief description of models

A literature search covering the decade 1982-1992 yielded seventeen publications pertaining to finite element models of the head which were produced during the period 1987-1992. In this section a global description of the various models is given following the grouping in Table 1.

2.1 Two-dimensional models

2.1.1 Two-dimensional human models

The major part of the reported studies is concerned with two-dimensional (2-D) models representing a sagittal section of a human or animal head. A finite element idealization of the mid-sagittal section of a human brain was constructed from a photograph by Lighthall *et al.* [1989]. With this model, consisting of a rigid skull and a homogeneous brain, Hybrid III sled tests were simulated using experimental acceleration-time histories of the center of gravity of the dummy head as input. The relative severity of the tests as judged by the HIC appeared to be opposite to that judged on the basis of either calculated maximum pressures or calculated maximum shear stresses. This model was also used in combination with a rhesus monkey model (M.-C. Lee *et al.* [1987]) to illustrate the application of a scaling law (Lighthall *et al.* [1989]; Ueno *et al.* [1989]). With the same model serving as a starting point Trosseille *et al.* [1992] carried out a parametric study to compare trends in simulation results with those of human cadaver experiments. For that end they extended the original model with an 'extremely rigid' beam representing the tentorium cerebelli. Furthermore, the influence of the no-slip condition at the skull-brain interface was diminished by incorporating a layer with a low shear modulus to model the cerebro-spinal fluid (CSF) surrounding the brain. In the original model brain compressibility was found to have a significant effect on the frequency response of the brain constrained by the skull; raising the value of the Poisson ratio from 0.49 to 0.499 resulted in an increase of the frequency of the first translational vibration mode from 195 Hz to 584 Hz. Adding a tentorium only slightly increased the frequencies of the first translational and rotational modes. The same applied to the incorporation of the CSF layer in the presence of the tentorium, as far as the first translational mode was concerned. However, adding the CSF layer in the configuration without tentorium caused this frequency to drop from 195 Hz to 83 Hz. A model of a parasagittal section of an average human head was developed by Chu and Lee [1991] to study mechanisms of cerebral contusions with special attention for coup-contrecoup phenomena. Validation against Nahum *et al.*'s [1977] experiments showed the calculated pressures to be roughly consistent with, but lower than, the experimental data. For frontal and occipital impacts almost identical tensile stresses were found in the contre-coup regions. The last human sagittal-plane model to be mentioned is that developed by Willinger *et al.* [1992] to study the coup-contrecoup mechanism. This model comprises a rigid skull, the brain, and a layer representing the subarachnoid space. In a modal analysis the Young's modulus of the subarachnoid layer was tuned so as to obtain a first rotational vibration mode of the brain at a frequency of 150 Hz, close to the value of 120 Hz found in experiments on human volunteers. The result was a virtually free boundary condition at the skull-brain interface realized by a very compliant subarachnoid layer. The responses to simulated frontal and occipital impacts showed a close resemblance if a time shift was applied.

Ruan *et al.* [1991a] conducted a study of side impact response using coronal-plane models of a 50th percentile human head. In addition they used an axisymmetric closed-sphere model to compare pressure distributions with those in the plane-strain models. A comparison was made between the response of the axisymmetric model and a human coronal model, both consisting of a deformable shell containing an inviscid fluid. The models were allowed to move freely in response to an impact impulse. In the axisymmetric model compressive stresses at the coup site and tensile stresses at the contrecoup site were observed, both having nearly equal magnitude. In the plane-strain model at both sites compressive stresses were found, those in the contrecoup region being smaller by a factor of 8 than in the coup region. These different behaviors were explained by the fact that the axisymmetric model is a closed system having a larger skull stiffness than the plane-strain model, which actually represents an open cylindrical shell. In the plane-strain model the influence on the pressure gradients of different boundary conditions at the head-neck junction was investigated which will be discussed in section 4. Finally the coronal model was extended by incorporating membranes to represent the dura, falx and tentorium as well as a CSF layer separating the skull, brain and membranes. The effect of the membranes was dramatic. Discontinuities in the pressure gradients at the membranes made the pressure distributions within the different brain compartments seemingly independent of each other. Moreover, from modal analyses the membranes revealed to have a significant stiffening effect on the brain, influencing notably the lower vibration modes.

2.1.2 Two-dimensional animal models

In the category of animal models only sagittal models were found. M.-C. Lee *et al.* [1987] simulated Abel *et al.*'s [1978] acceleration experiments with a model of a mid-sagittal section of a rhesus monkey to investigate mechanisms resulting in subdural hematoma. Translating their simulation results into terms of bridging vein deformations they found angular and translational accelerations to contribute equally to deformation for a Poisson ratio of 0.475. Decreasing the compressibility by taking a value of 0.49 reduced the contribution of the translational acceleration to about 40% of that due to angular acceleration. The possibly significant influence of the no-slip condition at the skull-brain interface on the results, by comparing them e.g. with those for a slip condition, was not investigated. Later this model was used to study the applicability of a scaling law (Lighthall *et al.* [1989]; Ueno *et al.* [1989]). With a model of the mid-sagittal brain section of a ferret, used in experiments, Ueno *et al.* [1991] simulated the GMRL controlled cortical contusion technique (Lighthall *et al.* [1989]). This technique consists of displacing the brain surface with a rigid surface impactor under controlled velocity and stroke. Fair agreement was observed between experimental and simulated pressure-time histories near the impactor. The model predicted high pressures not only in the impact region but also at sites remote from this region.

2.1.3 Two-dimensional physical models

Cheng *et al.* [1990] constructed a finite element model based on a half-cylinder physical model developed by Margulies [1987] as an idealization of a coronal brain section including a rigid falx. Before proceeding to a parametric study with this model a full-cylinder model was used to determine lacking values of some material constants for the surrogate brain material and to validate the modeling techniques against data from experiments with the corresponding physical model. The parametric study pertained to the influence on brain motion of skull-brain interface conditions, load magnitude and duration, and head size. In all cases maximum shear strains were found near the tip of the falx, which is in agreement with the findings of E.-S. Lee [1990], to be discussed in this section and Tong *et al.* [1989] (refer to section 2.2.2). A no-slip condition and a frictionless sliding interface condition were modeled. In the frictionless case an increase of the maximum shear strains was found, exceeding those in the no-slip situation by a factor of 2-11, depending on the load magnitude. Increasing the brain size led to an increase of the maximum shear strains; for small brain sizes the frictionless case showed larger strains than the no-slip case whereas for larger brains this effect was almost absent. The authors explained the difference between the responses of small and large brains in the frictionless case by pointing to the combined effects of size enlargement and application of a frictionless interface condition. The duration of the loading pulse affected the magnitude and the timing of the response but not its temporal characteristics.

E.-S. Lee [1990] developed 2-D finite element idealizations of full-cylinder, half-cylinder and skull physical models of Margulies [1987]. In a parametric study with the full-cylinder model with linearly elastic brain properties peak shear strains were seen to be linearly proportional to the magnitude of the imposed angular acceleration whereas with increasing model radius the strains increased following a parabolic curve. A similar parabolic relation was found in the half-cylinder model. The skull model with a no-slip condition at the skull-brain interface predicted maximum shear strains near this interface in the top and the bottom area. After extending the model with a rigid falx and a CSF layer between skull and brain, allowing relative brain-skull motion, maximum shear strains occurred near the falx tip. Adding lateral ventricles by assigning them CSF material properties led to a further increase of the shear strain values. Increasing the viscosity of the CSF by taking higher shear modulus values resulted in a decrease of the strain values.

2.2 Three-dimensional models

Relatively few 3-D models have been reported of late years. DiMasi *et al.* [1991a, 1991b] and Ruan *et al.* [1991b, 1992] constructed models of the human head. Mendis [1992], before proceeding to constructing a baboon and a human head model, developed 3-D models of the physical models of Margulies [1987] to validate his finite element modeling techniques. Tong *et al.* [1989] used a semi 3-D one-element thick 'slice' model based on a half-cylinder physical model of Margulies.

2.2.1 Three-dimensional human and animal models

The modeling efforts of DiMasi *et al.* [1991a, 1991b] are directed toward the estimation of strains induced in the brain in response to loading regimes representative of automobile collisions. First the modeling techniques were validated against data from laboratory crash tests with a Hybrid III dummy head (DiMasi *et al.* [1991a]). Next a 3-D head model was constructed comprising the skull, the brain with realistic coronal and sagittal geometries and the dura including the falx (Fig. 1). 25-mph impacts with a padded and

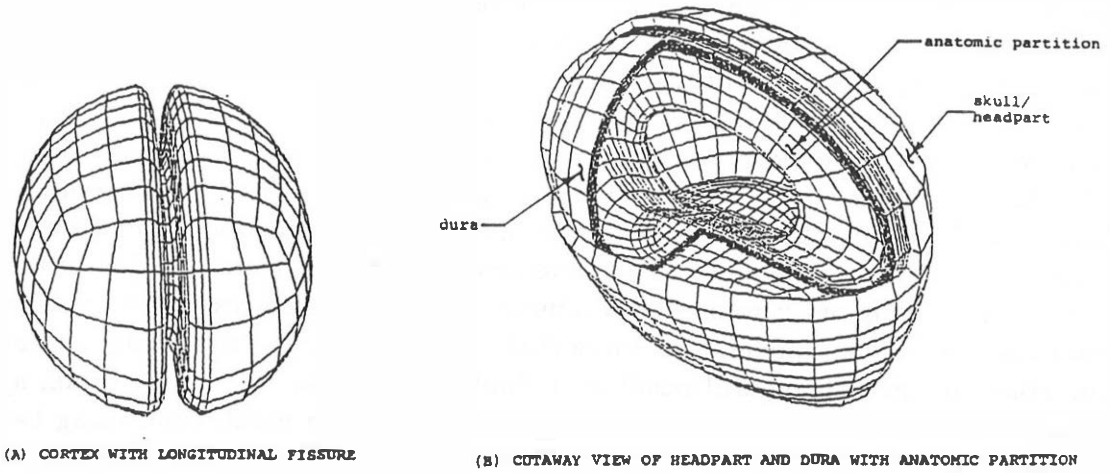

Figure 1: 3-D anatomic brain model (DiMasi *et al.* 1991b).

an unpadded automobile A-pillar were simulated (DiMasi *et al.* [1991b]). The A-pillars were modeled in detail. The head was launched into free flight with controlled speed and direction. By using contact algorithms a realistic simulation of the impact process was achieved. From a comparison of estimated centroidal accelerations with laboratory data it was concluded that the simulated dynamic loadings were representative for the automobile collision environment. Strains and strain rates for the unpadded condition were substantially larger than for the padded condition. The response for the unpadded case suggested traveling pressure wave effects whereas in the situation with padding coup-contrecoup phenomena were observed. Ruan *et al.* [1991b] studied pressure distributions in the brain in response to frontal impact using a model geometry based on Shugar's work [1977]. The model consisted of a three-layered skull and the brain with a CSF layer in between. The inertia of the facial structure was simulated by including the facial bone. The model was excited by use of an impact force-time history derived from Nahum *et al.*'s [1977] experiments. The pressure distributions agreed reasonably well with experimental data and showed the coup-contrecoup phenomenon. Shear stresses were three orders of magnitude lower than the pressure. Probably with the same model the same authors investigated the coup-contrecoup phenomenon in relation with the location of the impacting force (Ruan *et al.* [1992]). Fair agreement was found with the experimental data of Nahum *et al.* [1977]. The model predicted higher contrecoup pressures for rear impact than for frontal impact. For side impact coup and contrecoup pressures were virtually the same. Mendis [1992] developed 3-D models of a baboon and a human head to illustrate the feasibility of using finite element modeling techniques to extrapolate animal model rotational head injury data to estimate corresponding human injury thresholds. The baboon model was built up of a rigid skull containing brain, falx and tentorium. First the model was used to simulate experiments relating different lateral angular acceleration regimes to different grades of diffuse axonal injury (DAI). From the deformation responses of the model a strain measure was identified that correlated with the intensity of DAI. This strain measure, called oriented strain, was defined as the extension of infinitesimal line segments in the corpus callosum region oriented in the direction of the axonal bundles. In a parametric study the effect of translational acceleration on peak deformation was found to be negligible in comparison with the effect of rotational acceleration. A similar observation was made by M.-C. Lee [1987] with respect to deformation near the brain-skull interface in their 2-D rhesus monkey model (refer to section 2.1.2). Changes in model size and varying the stiffness of the brain had a significant influence on peak strains. Subsequently a double peak sinusoidal acceleration pulse equivalent with the exact experimental pulse was defined. A parametric study was done for different durations and magnitudes of this pulse. In a peak angular acceleration versus pulse duration plot isoparametric curves were constructed representing threshold estimates for different DAI levels. Next a human brain model was built, comprising the same substructures as the baboon model. Under the assumption that the human and baboon brain only differ in size but have similar shape and constitutive properties, the oriented strains in the same region as in the baboon were used to evaluate the deformation responses of parametric studies with the human model. Assuming that strain magnitudes that cause neurological dysfunctions in the baboon will result in a similar damage in the human brain, a set of isoparametric curves for the human was obtained. A crude verification using two case studies pertaining to actual human head accelerations gave satisfactory results as to the prediction of DAI intensity from these isoparametric curves.

2.2.2 Three-dimensional physical models

Mendis [1992] modeled and simulated a cylindrical physical model and a baboon skull physical model from the University of Pennsylvania (Margulies [1987]) to validate his finite element modeling technique. Tong *et al.* [1989] constructed a 3-D model of the same half-cylindrical UPenn physical model as used by Cheng *et al.* [1990] in a 2-D study. To elude the CPU-requirements of the 3-D model, however, they used only a one-element thick 'slice' of this model, representing a rigid cylinder with a rigid falx-like partition and the surrogate brain material (gel). The nodes at the gel-cylinder and the gel-falx interfaces were allowed to slide with free to choose friction levels. The effects on strain distribution of interface friction and the presence of the falx were investigated. The highest strain levels were found near the tip of the falx and appeared to be relatively insensitive to the level of friction. In the absence of a falx maximum principal strains occurred at the perimeter of the half-cylinder with no-slip conditions and more spatial variations for the zero friction case. This led to the conclusion that the influence on both magnitude and pattern of the induced strains of the falx is far more important than the interface friction.

A 3-D model of the Margulies full-cylinder physical model was used by E.-S. Lee [1990] to verify his finite element techniques and to obtain by trial and error constitutive parameter values so as to obtain deformations in his simulations that were similar to those in the experiments.

3 Constitutive properties

In all models discussed material properties were assumed to be homogeneous and isotropic. In the greater part of the studies linearly elastic constitutive models in combination with small-deformation theory is used, except in the work of Ueno *et al.* [1991] who employed large-deformation theory. In some cases structural damping was added (Table 2). Linearly viscoelastic properties combined with large-deformation theory is used in the brain model of DiMasi *et al.* [1991a, 1991b] and in all studies pertaining to the physical models of Margulies [1987]. In all these studies the standard linear solid or Flügge model is used in terms of the shear modulus

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t} \tag{1}$$

The material constants are defined and listed in Table 3. Mendis [1992], as the only exception, used non-linear stress-strain relationships for viscoelastic incompressible materials under large deformations. This constitutive model was obtained by utilizing a relaxation function form for the material coefficients and time-dependent strain invariants in a convolution integral form of the Mooney-Rivlin strain energy density function according to

$$U(t) = \int_{\tau=0}^{t} \{C_{10}(t-\tau)\frac{dI_1(\tau)}{d\tau} + C_{01}(t-\tau)\frac{dI_2(\tau)}{d\tau}\}d\tau \tag{2}$$

with strain invariants $I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2$; $I_2 = \lambda_1^2\lambda_2^2 + \lambda_2^2\lambda_3^2 + \lambda_3^2\lambda_1^2$, the λ 's representing the principal stretch ratios. This model was demonstrated to be suitable for modeling the large strain and high strain rate behavior of the Margulies full-cylinder silicone-gel physical model (Stalnaker and Mendis [1991]; Mendis [1992]). The brain and membrane properties that Mendis used in his baboon and human head models, were obtained by fitting the constitutive model to experimental data of Estes and McElhaney [1970] and Galford and McElhaney [1970], respectively, as:

brain: $C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-\frac{t}{0.008}} + 1103e^{-\frac{t}{-0.15}}$ (Pa)
membranes: $C_{01} = 0$; $C_{10}(t) = 6.72 \times 10^7\{1 - 0.816(1 - e^{-\frac{t}{0.001}})\}$ (Pa)

4 Boundary and interface conditions

Boundary and interface conditions constitute a very important feature of head impact models since they can have a significant influence on the model response. Reported modeling efforts in this respect relate to kinematic and dynamic boundary conditions at the skull-brain interface, dynamic boundary conditions at the foramen magnum and kinematic boundary conditions at the head-neck junction. Studies comparing the effects of different conditions at the skull-brain interface have been performed only with finite element idealizations of the half-cylinder physical models of Margulies [1987]. Galbraith and Tong [1989] utilized in a 3-D model different interface conditions ranging from pure slip via different friction levels to no slip. Higher friction levels induced higher strain levels. For the no-slip condition maximum principal strains were located near the boundary whereas for the slip condition more spatial variations were found. Tong *et al.* [1989] extended this model with a rigid falx and found the highest strain levels invariably near the tip of the falx. These large strains were relatively insensitive to the friction level although higher friction levels resulted in slightly lower strain levels as opposed to Galbraith's and Tong's configuration without a falx. The findings of Tong *et al.* [1989] were confirmed by the work of Cheng *et al.* [1990] who investigated in a similar, but 2-D model with a rigid falx the effects of no-slip and pure slip conditions (refer to section 2.1.3).

To these authors' knowledge no human head model studies explicitly addressing the effects of various skull-brain interface conditions have been reported in the past decade. DiMasi *et al.* [1991a, 1991b] applied slight friction (not quantified) to approximate slip conditions. A layer with a relatively low shear modulus, representing the CSF or the subarachnoid space, was used to allow relative brain-skull motion by Ruan *et al.* [1991b, 1992], Willinger *et al.* [1992] and, in a part of their simulations, by Trosseille *et al.* [1992]. From the latter two papers it is not clear whether the possibly large deformations of this layer are compatible with the small deformation theory which, very probably, has been used. In the papers by Ruan *et al.* no statement is made about the used deformation theory.

Lighthall *et al.* [1989], Ueno *et al.* [1989] and Trosseille *et al.* [1992] achieved a rigid-skull effect combined with a no-slip skull-brain interface condition by rigidly coupling the surface boundary nodes of the brain except at the foramen magnum so as to allow a force-free opening there. Chu and Lee [1991] used a deformable skull with a no-slip condition at the skull-brain interface. The foramen magnum was modeled as a force-free opening by defining two separate sets of nodes: one for the skull and one for the brain. They found the presence of the foramen magnum to have minor effects on the pressure distributions in the brain. Ruan *et al.* [1992] represented the foramen as a force-free opening by assigning the elements of the deformable skull at that site the material properties of the brain. Ueno *et al.* [1991] applied in their ferret model a pure-slip condition between the brain and the rigid skull while the foramen was modeled as a force-free opening.

Different kinematic boundary conditions at the head-neck junction were investigated by Ruan *et al.* [1991a] (Fig. 2). They used in their 2-D models a free boundary condition,

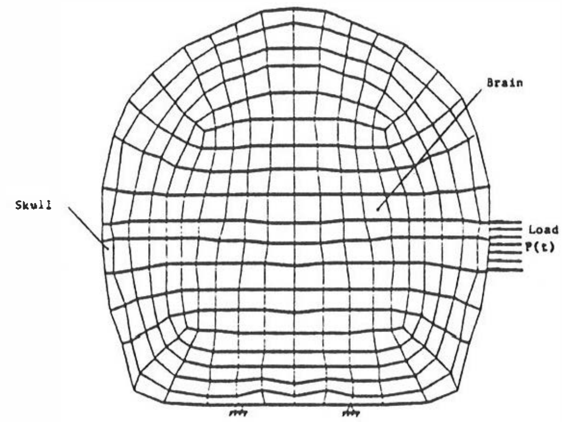

Figure 2: Plane strain model of human head, single layered shell filled with an inviscid fluid, double hinged (Ruan *et al.* 1991a).

a single hinge, and a double support, made up of a single hinge and a simple support. For the free boundary condition and the single hinge similar responses were obtained i.e. compressive stresses at the coup site and tensile stresses at the contrecoup site. The double support condition, however, changed the response dramatically by causing a transition of the coup stresses from compression to tension while the contrecoup stresses remained compressive. The authors concluded that this boundary condition, representing an extreme impact on an almost immobile head, imposes a too strong constraint and therefore should not be used in head injury modeling. Ruan *et al.* [1991b] and DiMasi *et al.* [1991a, 1991b] applied a free boundary condition at the head-neck junction under the assumption (for which DiMasi *et al.* put forward experimental evidence) that the neck is not likely to restrain the head during application of a short-duration impact pulse.

5 Discussion and concluding remarks

The state of the art with respect to constitutive models in head impact studies, as reflected by Tables 2 and 3, has not changed much since the early eighties. The material properties used have remained homogeneous, isotropic and linearly (visco)elastic. All but one of the elastic models were based upon small-deformation theory whereas in all viscoelastic studies large-deformation theory was used. The only significant step toward a more realistic representation of the material properties is constituted by Mendis' [1992] work, involving non-linear viscoelastic properties along with large deformations. Already in 1985 Thibault and Gennarelli stated: "... the development of sophisticated finite element codes promises to enable us to handle the non-linear aspects of geometry and material behavior. This also places an even more pressing burden on the experimentalist to provide detailed constitutive relations ...". Up to the present, however, the experimentalists did not take up the gauntlet. No new experimental data on intracranial tissue properties have been published since the early seventies. The data on hand lack consistency and completeness so that the quantification of model parameters from literature data is only possible by utilizing certain assumptions. Moreover, in default of proper descriptions of the macroscopic material properties, the results of isolated tissue studies (e.g. Galbraith *et al.* [1993]) will be hampered to show to full advantage.

In the past decade considerable attention was paid to the effect of different skull-brain interface conditions, but only relating to physical models. Establishing proper boundary conditions in the real head, not only at the skull-brain interface but also between the brain and the falx and tentorium partitions, remains a problem yet to be solved. The same is true for the kinematic and dynamic boundary conditions at the head-neck junction. As to the kinematic boundary condition it can be stated that -although Ruan *et al.*'s [1991a] work seems to point to an equivalence of a free and a single hinge condition-Khalil and Viano's [1982] conclusion still holds: the most reasonable boundary condition probably depends on the impact condition and obviously needs further investigation. Further research is also needed with respect to the dynamic boundary conditions at the foramen magnum. Because of the presence of openings, notably the foramen magnum, the skull-brain system is not a closed system. The pressure in the brain may change not only due to motion of brain tissue through the foramen but also through change in volume of the CSF-filled ventricles and the vasculature bed along with inflow or outflow of CSF and blood during impact. Therefore, although it is widely endorsed that it is quite reasonable to model brain tissue as a nearly incompressible material, the skull-brain system is somewhat compressible and often an effective compressiblity is used. It seems worthwhile to undertake efforts directed toward more realistic models by explicitly taking into account the effects of the presence of the foramen, the ventricles and the vasculature. Representing the foramen by a force-free opening in combination with a no-slip skull-brain interface condition (Lighthall *et al.* [1989]; Ueno *et al.* [1989]; Troseille *et al.* [1992]; Chu and Lee [1991]) is likely to impose a too strong constraint on movement of brain material through the spinal canal opening. Using a slip condition at the skull-brain interface along with assigning to the foramen the impedance of the spinal canal might be more appropriate. Considerable problems relating to fluid-structure interaction may be expected when one attempts to model the brain vasculature and the ventricles. Starting points to alleviate these problems may be found in the field of multi-phasic material models (e.g. Oomens *et al.* [1987]).

Making a qualitative, let alone a quantative, comparison between the various studies is far from straightforward. Not only models with different geometries but also models with similar geometries but different sizes will respond differently to the same loading conditions. Mendis [1992] found moderate changes in size of his 3-D baboon model ($\pm$ 10% of the nominal size) to have a very strong effect on peak strains. Lee [1990] and Cheng *et al.* [1990] investigated the influence of model size in 2-D half-cylinder models (without and with a rigid falx, respectively). Lee, using a no-slip condition at the skull-brain interface, observed peak shear strains to increase with increasing model size, both for linearly elastic and viscoelastic surrogate brain material properties. The strain versus model radius curves resembled parabolic functions which is in agreement with Holbourn's scaling method. This method also correlated fairly well with Cheng *et al.*'s results for a no-slip boundary condition at the skull-brain interface but gave an over-prediction of peak shear strains by a factor of 2.5 for the frictionless condition. Holbourn's method is based on the mass ratio of the models. Ueno *et al.* [1989] and Lighthall *et al.* [1989] used a scaling method based on the area ratio of their 2-D models which in contrast to Holbourn's method, also involved time-scaling. Scaling mechanical parameters from a monkey model to a human model resulted in similar overall characteristics of the stress distributions. However, the maximum shear stress in the human model exceeded that in the monkey model by a factor of 1.7. Clearly, application of scaling methods not only to different geometries but even to similarly-shaped brain models with different boundary conditions needs further investigation.

Apart from Ueno *et al.*'s [1991] paper most of the publications provide no or insufficient information concerning the used space-time discretizations as well as numerical integration techniques enabling an evaluation with respect to both overall accuracy and the suitability of the models to capture wave propagation phenomena.

The value of physical models is underlined by several researchers who utilize the physical models of Margulies [1987], first to validate their finite element modeling techniques and subsequently as a starting point for parametric studies. The sophistication and complexity of finite element models will undoubtedly increase in the near future. Reproducible experiments with physical models will be indispensable for a first validation of these models. The design of the physical models should be such that parameters -especially those relating to material properties and interface boundary conditions- can be manipulated in a controlled manner.

The necessity of three-dimensional modeling versus the suitability of two-dimensional analyses is still a controversial issue. In the reviewed period four different 2-D human models and three different 3-D human models have been reported (Table 1). 3-D simulations are time-consuming and thus expensive, as was illustrated by DiMasi *et al.* [1991a] who needed 2 hours of CPU time on a CRAY2 computer to simulate a 3-D impact event during 20 ms. According to Cheng *et al.* [1990] 2-D analyses are more cost effective for studying effects of different model parameters and boundary conditions. Once these effects have been determined, the influence of skull geometry must be studied in 3-D analyses. An example in support of this point of view is the fact that Cheng *et al.* [1990] in a 2-D study arrive at conclusions, relating to the effect of skull-brain interface conditions, that are in general agreement with those of Tong *et al.* [1989] for a 3-D version of the same model. However, counter-examples can also be found. Ruan *et al.* investigated coup-contrecoup pressures in response to side impact with different models. In an axisymmetric closed-skull model (Ruan *et al.* [1991a]) and in a 3-D model (Ruan *et al.* [1992]) they observed similar pressure variation patterns i.e. compressive stresses at the coup site and tensile stresses at the contrecoup site. A 2-D model (Ruan *et al.* [1991a]) on the other hand yielded compressive stresses at the contrecoup site, albeit by a factor of 8 smaller than at the coup site. Another example relates to the occurrence of coup-contrecoup lesions due to frontal and occipital impacts. Chu and Lee's [1991] 2-D model predicted equal chance for frontal and occipital impacts to cause coup and contrecoup lesions. Ruan *et al.* [1992] found in their 3-D model higher contrecoup pressures due to rear impact which is in agreement with the clinical observation that contrecoup injury occurs more frequently in the frontal lobe. The latter examples at least suggest that one should be very cautious when extrapolating results from 2-D analyses to 3-D situations. Especially under large deformations plane strain models may provide unrealistic results because they will maintain under all circumstances the plane character of the strain distributions whereas these in reality are likely to be of an essentially three-dimensional nature. The developments with regard to the power of computers as well as the capabilities of comercially available finite element codes (used in each study discussed in this paper) have been enormous. Consequently the necessity to restrict modeling efforts to two-dimensional idealizations is expected to become less compelling in the near future.

Taking into account that the reviewed period covers a decade with considerable developments of hardware and software, one cannot but conclude that progress in finite element modeling of head impact has been rather slow. The existing finite element models are still far from explaining injury mechanisms and predicting type and severity of injuries in relation to loading conditions.

References

Abel, J.M., T.A. Gennarelli and H. Segawa. Incidence and severity of cerebral concussion in the rhesus monkey following sagittal plane angular acceleration. *Proc. 22nd Stapp Car Crash Conf.*, 1978, p. 35-53.

Cheng, L.Y., S. Rifai, T. Khatua and R.L. Piziali. Finite element analysis of diffuse axonal injury. *Proc. 34th Stapp Car Crash Conf.*, SAE Paper 900547, 1990, p. 141-154.

Chu, C.-S. and M.-C. Lee. Finite element analysis of cerebral contusion. *Advances in Bioengineering*, ASME-BED-Vol. 20, 1991, p. 601-604.

DiMasi, F., R.H. Eppinger, H.C. Gabler III and J.H. Marcus. Simulated head impacts with upper interior structures using rigid and anatomic brain models. *Auto & Traffic Safety.* Summer 1991a, p. 20-31.

DiMasi, F., J. Marcus, R. Eppinger. 3-D anatomic brain model for relating cortical strains to automobile crash loading. *Proc. 13th Int. Techn. Conf. on Experimental Safety Vehicles*, November 4-7 1991b, Paper no. 91-S8-O-11.

Estes, M.S. and J.H. McElhaney. Response of brain tissue to compressive loading. ASME-Paper no. 70-BHF-13, 1970.

Galbraith, C.G. and P. Tong. Boundary conditions in head injury finite element modeling. *16th Annual Int. Workshop on Human Subjects for Biomechanical Research*, 1988, p. 179-193.

Galbraith, J.A., L.E. Thibault and D.R. Matteson. Mechanical and electrical responses of the squid giant axon to simple elongation. *J. Biomechanical Engineering*, Vol. 115, 1993, p. 13-22.

Galford, J.E. and J.H. McElhaney. A viscoelastic study of scalp, brain and dura. *J. Biomechanics*, Vol. 3, 1970, p. 211-221.

Gennarelli, T.A., L.E. Thibault, J. Adams, D. Graham, C. Thompson and R. Marcinin. Diffuse axonal injury and prolonged coma in the primate. *Ann. Neurology*, Vol. 12, 1982, p. 564-574.

Khalil, T.B. and D.C. Viano. Critical issues in finite element modeling of head impact. *Proc. 26th Stapp Car Crash Conf.*, SAE Paper 821150, 1982, p. 87-102.

Lee, E.-S. A large-strain, transient-dynamic analysis of head injury problems by the finite element method. Ph.D. Dissertation, Georgia Institute of Technology, 1990.

Lee, M.-C., J.W. Melvin and K. Ueno. Finite element analysis of traumatic subdural hematoma. *Proc. 31st Stapp Car Crash Conf.*, SAE Paper 872201, 1987, p. 67-77.

Lighthall, J.W., J.W. Melvin and K. Ueno. Toward a biomechanical criterion for functional brain injury. *Proc. 12th Int. Techn. Conf. on Experimental Safety Vehicles*, 1989, p. 627-633.

Margulies, S.S. Biomechanics of traumatic coma in the primate. Ph.D. Dissertation, University of Pennsylvania, 1987.

Mendis, K. Finite element modeling of the brain to establish diffuse axonal injury criteria. Ph.D. Dissertation, Ohio State University, 1992.

Nahum, A.M., R. Smith and C.C. Ward. Intracranial pressure dynamics during head impact. *Proc. 21st Stapp Car Crash Conf.*, SAE Paper 770922, 1977, p. 339-366.

Oomens, C.J.W., D.H. van Campen and H.J. Grootenboer. A mixture approach to the mechanics of skin. *J. Biomechanics*, Vol. 20, No. 9, 1987, p. 877-885.

Ruan, J.S., T. Khalil and A.I. King. Human head dynamic response to side impact by finite element modeling. *J. Biomechanical Engineering*, Vol. 113, 1991a, p. 276-283.

Ruan, J.S., T. Khalil and A.I. King. Intracranial response of a three-dimensional human head finite element model. *Proc. Injury Prevention through Biomechanics Symposium*, Wayne State University, 1991b, p. 97-103.

Ruan, J.S., T. Khalil and A.I. King. Finite element analysis of the human head to impact. *Advances in Bioengineering*, ASME-BED-Vol. 22, 1992, p. 249-252.

Shugar, T.A. A finite element head injury model. Technical report TR-R-854-1, Vol. 1, Civil Engineering Laboratory, Naval Construction Battalion Center, Port Hueneme, California 93043, 1977.

Stalnaker, R.L. and K.K. Mendis. State of the art of rotational head injury modeling. *Proc. Injury Prevention through Biomechanics Symposium*, Wayne State University, 1991, p. 105-112.

Thibault, L.E. and A. Gennarelli. Biomechanics and craniocerebral trauma. In: *Central Nervous System Trauma Status Report*, Eds.: D.P. Becker and J.T. Povlishock. NINCDS, NIH, 1985, p. 379-389.

Tong, DiMasi, Carr, Eppinger, Marcus and Galbraith. Finite element modeling of head injury caused by inertial loading. *Proc. 12th Int. Conf. on Experimental Safety Vehicles*, 1989, p. 617-626.

Trosseille, X., C. Tarrière, F. Lavaste, F. Guillon and A. Domont. Development of a F.E.M. of the human head according to a specific test protocol. *Proc. 36th Stapp Car Crash Conf.*, SAE Paper 922527, 1992, p. 235-253.

Ueno, K., J.W. Melvin, E. Lundquist and M.C. Lee. Two-dimensional finite element analysis of human brain impact responses: Application of a scaling law. *Crashworthiness and Occupant Protection in Transportation Systems*, AMD-Vol. 106/BED-Vol. 13, Ed.:T.B. Khalil, ASME, 1989, p. 123-124.

Ueno, K., J.W. Melvin, M.E. Rouhana and J.W. Lighthall. Two-dimensional finite element model of the cortical impact method for mechanical brain injury. *Crashworthiness and Occupant Protection in Transportation Systems*, AMD-Vol. 126/BED-Vol. 19, ASME, 1991, p. 121-147.

Willinger, R., C.M. Kopp and D. Césari. New concept of contrecoup lesion mechanism: modal analysis of a finite element model of the head. *Proc. Int. IRCOBI Conf. on the Biomechanics of Impacts*, 1992, p. 283-297.

Table 1: General Description of 2-D and 3-D models

			Author	Specification	Substructures	Input	Injury	Validated against	Code[3]
2-D Model	human	sagittal	Lighthall [1989]	[1]	rigid skull, brain, foramen	acceleration Hybrid III experiment			MSC/NASTRAN (S)
			Ueno [1989]	[1]	rigid skull, brain, foramen	rotational acceleration			"
			Chu [1991]	'average' human head	skull, brain	displacement	contusion coup-contrecoup	human cadaver exp., Nahum [1977]	ANSYS (S)
			Trosseille [1992]	[1]	rigid skull, brain, foramen, CSF, rigid tentorium	modal analysis		human cadaver exp.; accel. measurements on boxers	MSC/NASTRAN (S)
			Willinger [1992]	'a' human head	rigid skull, brain, subarachnoid space	modal analysis & impact force	coup-contrecoup	in vivo & in vitro vibration analysis of human heads	
		5	Ruan [1991a]	50th percentile human head	skull, brain, dura, falx, tentorium, CSF	impact force			MARC, ANSYS (S)
	animal	sagittal	Lee [1987] Lighthall [1989] Ueno [1989]	rhesus monkey	rigid skull, brain	idealized acceleration of Abel [1978]	subdural hematoma		MSC/NASTRAN (S)
			Ueno [1991]	ferret	rigid skull, brain	acceleration		controlled cortical impact experiments	ABAQUS (L)
	physical [4]		Cheng [1990]	full- and half-cylinder model	rigid cylinder, rigid falx, gel (surrogate brain)	velocity	DAI	UPenn[2] experiments	DYNA2D (L)
			Lee [1990]	full- and half-cylinder model; skull model	rigid cylinder, rigid falx, gel (surrogate brain), CSF, ventricles	displacements of perimeter nodes	DAI	UPenn[2] experiments	PATRAN (L)
3-D Model	human		DiMasi [1991a,1991b]		skull, brain, dura,falx	velocity			DYNA3D (L)
			Ruan [1991b]	based on Shugar's [1977] model of 50th percentile head	3-layered skull, brain, CSF, inertia facial bone	impact force Nahum [1977]		human cadaver exp.; Nahum [1977]	PAMCRASH (?)
			Ruan [1992]	"	" + foramen	"		"	"
			Mendis [1992]		rigid skull, brain, falx, tentorium		DAI		
	6		Mendis [1992]	baboon	rigid skull, brain, falx, tentorium	acceleration	DAI	experiments of Gennarelli et al. [1982]	ABAQUS (L)
	physical [4]		Galbraith [1988]	half-cylinder	rigid cylinder, gel (brain)	exp. load-time history		UPenn[2] experiments	DYNA3D (L)
			Tong [1989]	"	rigid cylinder, gel (brain), rigid falx	actual motion of perimeter nodes		"	"
			Lee [1990]	full-cylinder	rigid cylinder, gel (brain)	"		"	PATRAN (L)
			Mendis [1992]	"	"	acceleration		"	ABAQUS (L)

[1] same model
[2] UPenn: University of Pennsylvania
[3] (S)/(L): small/large deformation
[4] physical models of Margulies [1987]
[5] coronal
[6] animal

Table 2: Mechanical properties of tissues in the linearly elastic models.
E: Young's modulus; G: shear modulus; K: bulk modulus; ρ: density;
υ: Poisson ratio; tan(δ): loss tangent.

Author	Substructures[1]	E (kPa)	G (kPa)	K (kPa)	ρ (kg/m³)	υ	tan(δ)
Lee [1987] Lighthall [1989] Ueno [1989,1991]			80.0		1000	0.475 & 0.49	0.2
Lee [1990]	brain (gel) [2]	80.0-121.2				0.49	
Chu [1991]	skull	6.5×10^6			2027	0.2	0.001
	brain	250.0			1000	0.49	0.001
Ruan [1991a]	skull	6.5×10^6			1412	0.22	
	brain	66.7		2.19×10^2 - 2.19×10^6	1040	0.48 - 0.49999492	
	dura, falx, tentorium	31.5×10^3			1133	0.45	
	CSF	66.7			1040	0.499	
Ruan [1991b]	outer table		5.0×10^6	7.3×10^6	3000	0.22	
	diploe		2.32×10^6	3.4×10^6	1750	0.22	
	inner table		5.0×10^6	7.3×10^6	3000	0.22	
	brain		1.68×10^3	2.19×10^3	1040	0.4996	
	CSF		500	2.19×10^4	1040	0.489	
Ruan[3] [1992]							
Trosseille [1992]	brain	240.0			1000	0.49 - 0.499	0.2
	CSF		0.2			0.49999	
Willinger [1992]	skull	5×10^6				0.2	
	brain	675.0				0.48	
	subarachnoid space	5×10^{-2}				0.49	

[1] skull is not mentioned if rigid
[2] in 2-D and 3-D full-cylinder model
[3] no values given, probably identical to Ruan [1991b]

Table 3: Mechanical properties of tissues in the linear viscoelastic models.
E: Young's modulus; G_∞: static shear modulus; G_0: dynamic shear modulus;
β: decay constant; ρ: density; υ: Poisson ratio; K: bulk modulus.

Author	Substructures	E (kPa)	G_∞ (kPa)	G_0 (kPa)	β (s^{-1})	ρ (kg/m³)	υ	K (MPa)
Galbraith [1988] Tong [1989]	brain (gel)		5.512	11.02	200		0.4995	
Cheng [1990]	brain (gel)		16.2	49.0	145		~ 0.5	
Lee [1990]	brain (gel) [1]		2.87 -18	26.9 -110	50	950		1.25 -5.44
	CSF/ventricles [2]		3 - 6	9 - 24	50			0.445
DiMasi [1991a, 1991b]	skull	2.4×10^6						
	brain		17.225	34.45	100			0.0689
	dura	6890						

[1] used in half-cylinder, full-cylinder and skull physical models of Margulies [1987]
[2] used in skull physical models of Margulies [1987]