Finite Element Simulation of Deformation Characteristics of the Human Skull

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ABSTRACT

To analyse and quantify the injury process of the human head and to develop enhanced injury sensing criteria for automotive crashes an anatomical three-dimensional finite element model of a human head was developed. The detailed skull geometry based on Computer Tomography scans. This paper outlines results from a joint research program at the Institute of Automotive Engineering of the Technical University Berlin and the Institute of Forensic Medicine of the University Heidelberg. To integrate reliable material properties a series of tests was carried out in Heidelberg to determine basic mechanical properties of isolated human cranial bones using fresh bone specimens. The results of these experiments have been analysed in terms of statistical parameters to describe the characteristics of these materials. Experimental test results were compared to finite element simulation to determine and describe validated numerical input parameters for the finite element head model.

INTRODUCTION

Numerical occupant simulation models based on rigid multi body systems of current Dummy models. Finite element modelling efforts are concerned with finite element models of rigid Dummy models. The "third generation" of finite element Dummy models is partly or completely modelled with deformable finite elements, which lead to potentially high precision response for a much wider range of loading situations /10/. It is obvious that the impact response of a human body is completely different from a rigid body response. Therefore models of parts of the human body are needed to obtain more realistic information about their response in crash events. The human head is of prime importance in the development of these models because of its frequent involvement in fatal injuries.

Our approach is to use a detailed finite element model of the human head, to study the relationship between input parameters and corresponding local output parameters and to obtain more information about the injury mechanisms and the response for dynamic loading. Finite element analysis seems to be the most appropriate analytical technique for analysing the geometrical and mechanical complex human structures. The finite element method provides reliable and reproducible results from adequate geometry and material models.

Limitations of finite element modelling efforts in head injury were questioned in early 1980's caused by the assumptions and approximations made at the time /4/. Most of the assumptions were necessary due to technical limitations of the finite element method of the time. A recent review of FE modelling of head impact was published last year /9/.

The finite element simulation is generally subdivided into three different parts, figure 1:

- development of geometrical models and determination of material properties and boundary conditions (pre-processing)
- numerical solution with the finite element program code (main processing)
- graphical and mathematical interpretation of results (post processing)

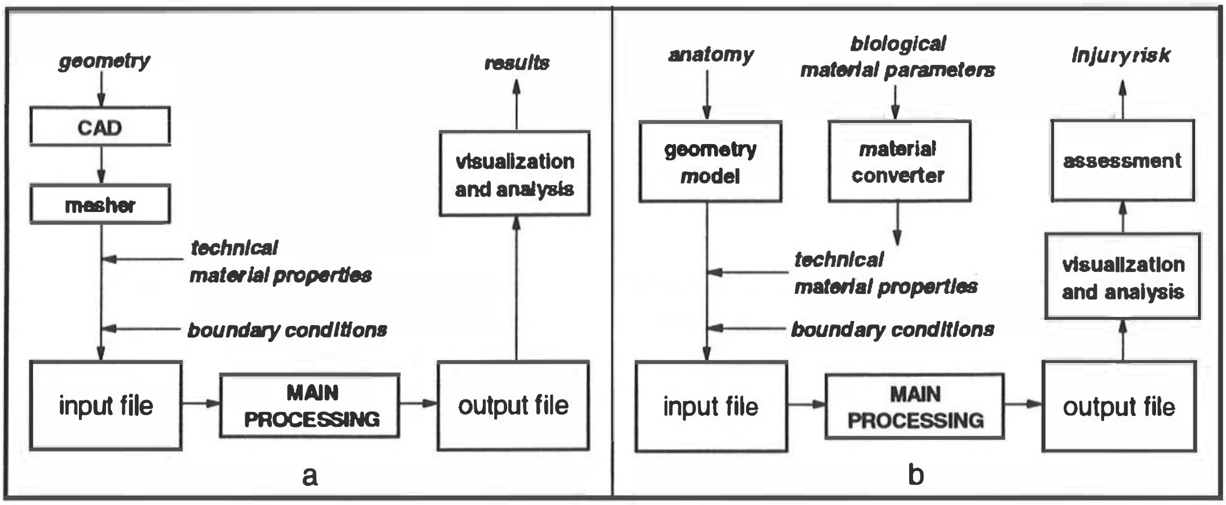

Figure 1. *Finite element modelling process in the automotive development (left) and biomechanics (right)*

In the automotive development CAD data were converted into finite element data by mesh generator programs. The material data and the boundary conditions are normally well known and described in the automotive environment (figure 1a). The application of finite element modelling to biomechanics requires different methods to design a model with accurate geometry and material properties. Normally geometry data files from most of the human parts are hardly available. Furthermore biological material properties must be converted into suitable numerical data to describe the mechanical behaviour of the model. For the post-processing part it is necessary to analyse complex output parameters and to assign the injury risk to physical units, such as displacements, stress or pressure (figure 1b).

The aim of our research work described in this paper is to provide a basis of material properties in sufficient detail for the human head model. The model will allow systematic study of various head injury mechanisms. The code used for the simulations is the non-linear explicit finite element code PAM-CRASH which is widely used in the automotive industry for dynamic crash simulation /1/.

HUMAN HEAD MODEL

The three-dimensional human head model that represents the exact anatomy of a human head is shown in Figure 2. Computer Tomography scans were transformed with a self-developed pre-processor program into a PAM-CRASH input file. The model consists of 1.342 solid elements with 2.874 nodes in 31 layers and represents the bony structures of a human skull without the lower jaw /5/.

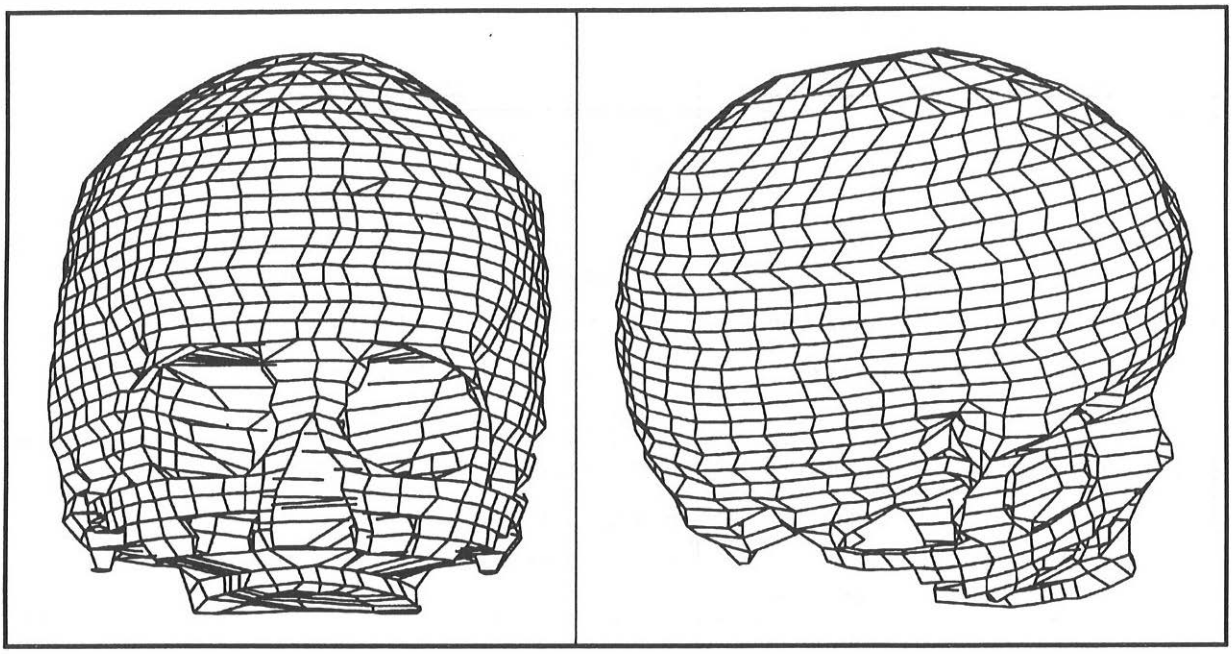

Figure 2. Exterior frontal and lateral view of the anatomical finite element model of the human skull

MECHANICAL PROPERTIES

Although skull fractures were studied for years the literature on the mechanical properties of the skull is quite modest. Tests were carried out to determine the material properties of the skull and the brain, but biological parameters were insufficiently described by analytical material models. Several studies compared experimental data with results from accident investigation research. These data confirm that head injuries rank as a major cause of death and disability in the world but provide not the necessary biomechanical input parameters /2/.

Different experiments were determined to study the mechanical properties and behaviour of skull bone. In such studies static tensile and compressive strengths of embalmed human parietal bone or the mechanical properties of unembalmed human compact cranial bone in tension were reported /3, 6, 8/. For the development of a mechanical model of the human head the mechanical behaviour of the diploë layer in compression was investigated /7/.

Table 1 shows exemplary mechanical properties of human skull bone specimens. The wide range of different material properties resulted from dissimilar test conditions and various biological parameters of the individual human specimen. The test results found in the literature do not fulfil the requirements for a finite element material. Information about the specimens, the test procedures and the test devices in some cases is missing or incompletely described.

The objectives of the finite element modelling were the reason for conducting material tests at the Institute of Forensic Medicine (Heidelberg) in co-operation with the Institute of Automotive Engineering (Berlin). The tests were performed with isolated fresh bone specimens on a universal tensile test-machine. The test series included quasistatic compression test and three-point bending test where force deflection curves were measured /12/.

Table 1: Mechanical properties of human skull bones

Author	Specimen	Material properties
McElhaney et al. /6/	Cranium	Compression radial: Young modulus: E = 241,2 N/mm^2; Poisson ν = 0,19; tangential: E = 558,4 N/mm^2; ν = 0,22
Melvin et al. /7/	Compact bone	Young's modulus (tension): E = 12.41-19.99 N/mm^2; Fracture stress: σ_B = 68,9 - 96,46 N/mm^2
	Spongy bone	Young's modulus: E = 1.378 N/mm^2
Wood et al. /14/	Cranium	Young's modulus tension: E = 10.335 - 22.048 N/mm^2
		Fracture stress: 48,2 - 127,5 N/mm^2
Schneider et al. /11/	Mandible	Fracture force: A-P 1.780 N; lateral 890 N
	Maxilla	Fracture force: 668 N
	Os zygomaticum	Fracture force: 890 N
Simkin et al. /13/	Skull	Young's modulus bending: E = 5.984 N/mm^2

METHOD

The bones of the cranium each have their own characteristic structural variations. Their basic structure is that of an inner and outer layer of compact bone separated by a layer of porous bone known as the diploë layer. The aim of the research was to specify the mechanical properties of the skull bones relevant to the finite element model of the human head. The input data for the finite element simulation were the specimen geometry, the force deflection curves and the test conditions from the Heidelberg tests, figure 3. The numerical model combines the material properties of both compact layer and diploë in one material.

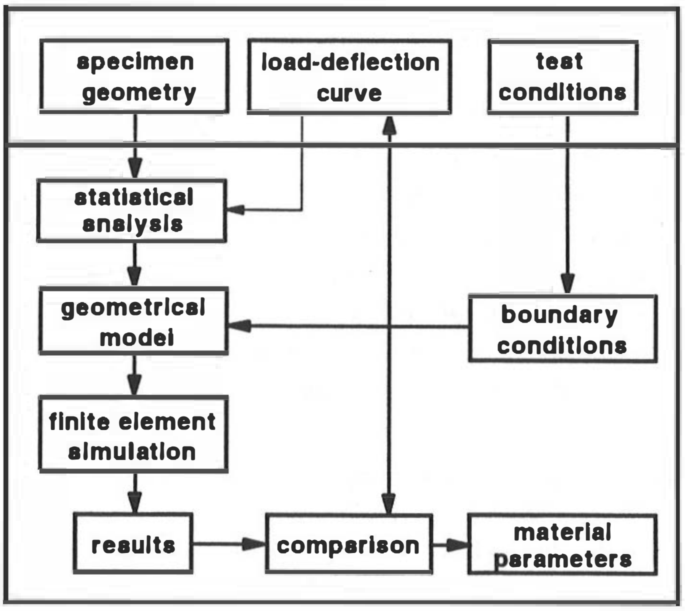

Figure 3. Determination of material characteristics from experimental tests

Parameters not available from the experiments (damping-factor, friction-factor of aluminium or bones) were determined and estimated. The influence and effects of additional parameters like number, shape of the elements, defined contact zones (sliding interfaces), mechanical properties and program-specific values were investigated with the PAM-CRASH program code. To validate the numerical material model the experimental measured and analytical calculated force-deflection curves were compared.

STATISTICAL ANALYSIS

The great variety of force deflection curves from the bending and compression tests (figure 4) required a statistical analysis of the anatomical and geometrical data and the measured force deflection responses.

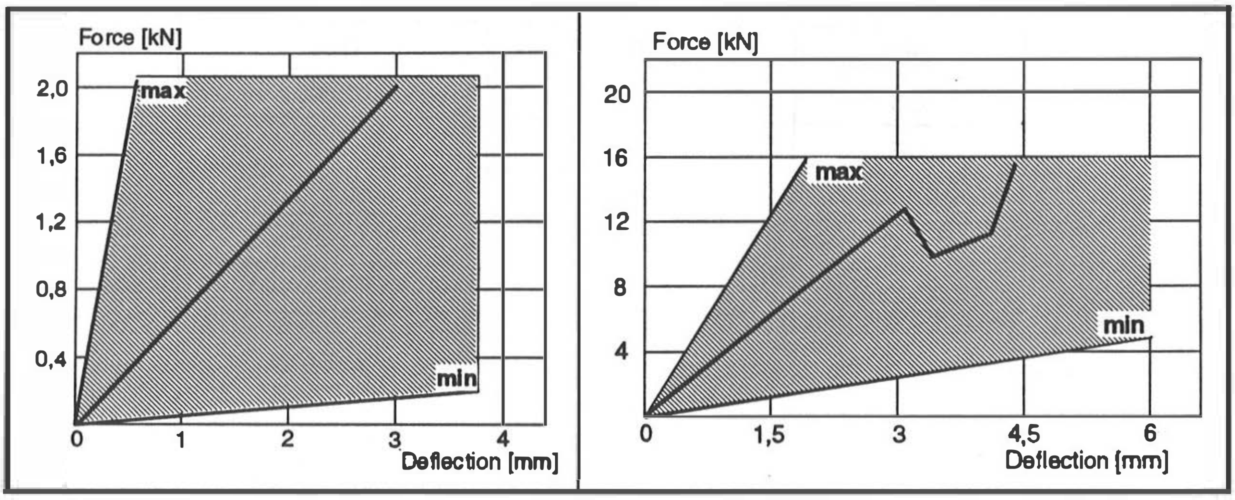

Figure 4. Variety of force deflection curves for human skull bone specimens in bending (left) and compression (right)

In addition to the physiological parameter age and sex, geometrical and histological parameters for every test specimen were documented, table 2.

Table 2 Summary of measured test parameters of human skull bone specimen

geometrical parameters	histological parameters
thickness compacta extema (DKE)	water content compacta (WGK), diploë (WGS)
thickness compacta intema (DKI)	content of mineral substances compacta (MGK), diploë (MGS)
thickness diploë (DS)	content of organic substances compacta (OSK), diploë (OSS)
total thickness (GD)	density

Geometrical parameters: To reduce the individual varieties of the Heidelberg test data the proportions of the thickness of compacta intema, extema and diploë of all specimens were correlated with each other. To compare the thickness of the layers the relative thickness of the specimens related to the total thickness was defined. Table 3 shows a two-tailed Pearson correlation analysis at a level of significance $p<0.001$ and describes the mean values and standard deviations of the geometrical parameters.

Table 3 Correlation analysis: relative thickness of layered cranial bone (parietal); mean values and standard deviations of the geometrical parameters (n = 284)

parameters	DKE/GD	DKI/GD	DS/GD
DKE/GD		r = 0.72	r = 0.89
DKI/GD	r = 0.72		r = 0.92
DS/GD	r = 0.89	r = 0.92	
r = Pearson correlation coefficient; number of cases n = 284			
parameters	mean	standard deviation	
DKE	1.60	0.47	
DKI	1.37	0.51	
DS	3.15	1.22	
GD	6.11	1.19	
DKE/GD	0.265	0.07	
DKI/GD	0.225	0.08	
DS/GD	0.510	0.14	

The regression curves of the relative thickness and the density were compared to determine the range of density of layered cranial bone. The density ranges from 1.62 to 1.71 kg/m^3. The mean density of all specimen is 1.66 kg/m^3 with a standard deviation of 0,140. The correlation analysis of the geometrical parameters shows an average distribution of the relative thickness of layered cranial bone as outlined in figure 5.

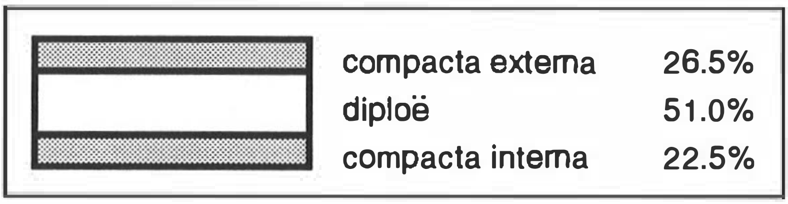

Figure 5. Average relative thickness of human cranial bone (parietal)

Mechanical parameters: The correlation analysis of the geometrical, physiological and histological parameters showed numerous interactions. Remarkable are the good correlations between the histological parameters and the relative thickness of compacta and diploë. Further no or minor interactions with physiological parameters were found.

The next task was to analyse the interactions with the mechanical parameters to enable the prediction of the mechanical properties with only a few measurable biological parameters of the bone. For this purpose the force-deflection curves of the compression tests were divided into linear areas. Figure 6 shows the typical sectors of the compression and bending force deflection curves.

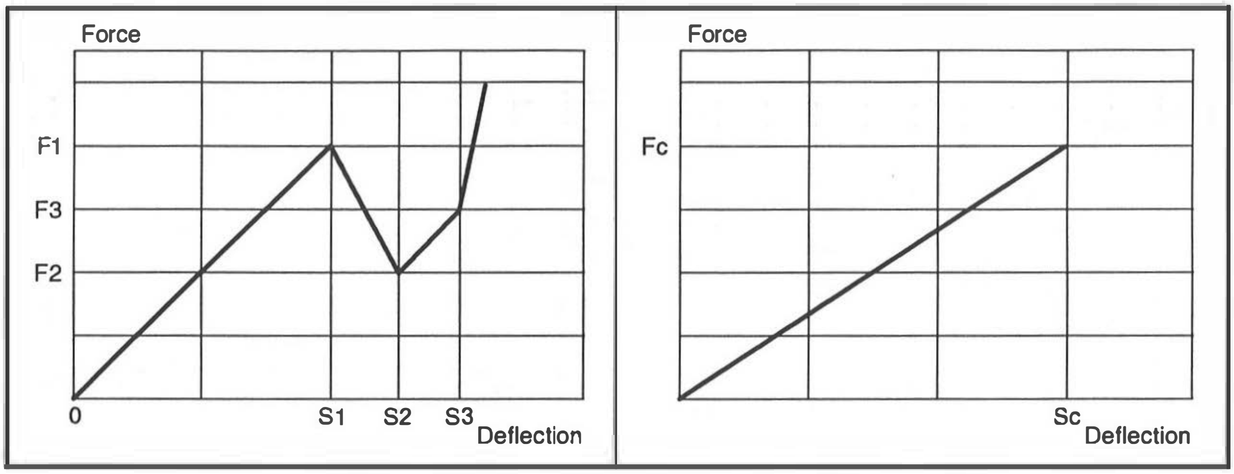

Figure 6. Typical force deflection curve for human skull bone in compression and bending

Compression tests: The deflection S1 until the diploë crush correlates well with density, thickness of diploë (DS) and the relative thickness of compacta extema, intema (DKE/GD, DKI/GD) and diploë (DS/GD), figure 7a. The failure force F1 of the diploë correlates with thickness of diploë (DS), figure 7b. The analysis shows no significant correlation between mechanical compression and physiological or histological parameters.

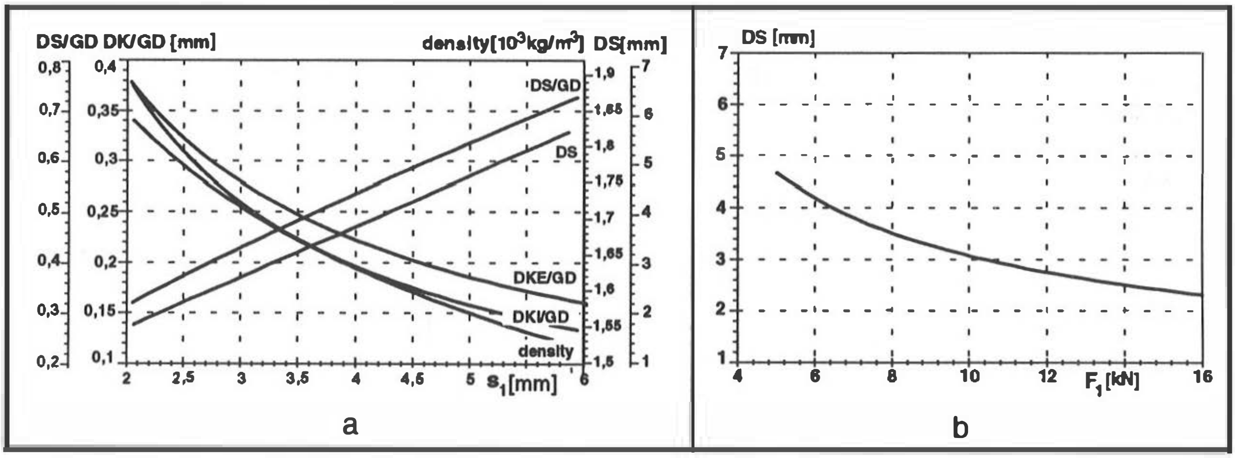

Figure 7. Regression curves of mechanical parameters and significant influence parameters for bone in compression (regression type with the lowest sum of squared errors)

Bending tests: The failure force F_{crush} correlates with age, total thickness and the thickness of compacta intema and extema (DKE, DKI), figure 8a. The failure deflection s_{crush} correlates with age and total thickness of the specimen, figure 8b. The analysis shows no significant correlation between mechanical bending parameters and histological parameters.

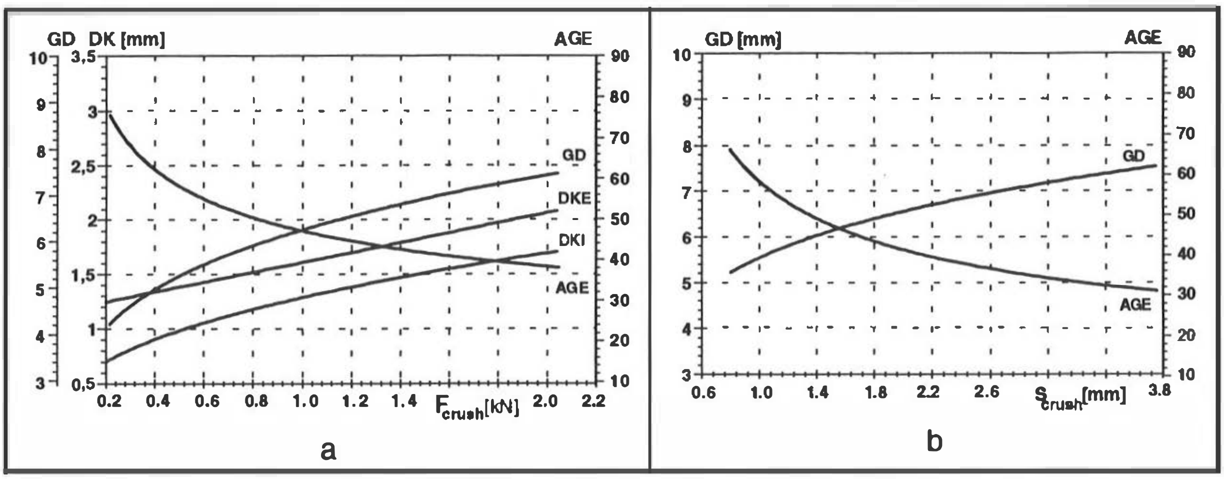

Figure 8. Regression curves of mechanical parameters and significant influence parameters for bone in bending

FINITE ELEMENT SIMULATION

Compression tests: Four representative tests were chosen for the numerical simulation. The square specimen had a length and width of 10 mm, the height ranged from 3,52 to 9,78 mm with an average height of 6,22 mm. The model consists of 1.000 elastic-plastic brick elements with elastic-plastic material type. The finite element model of the test configuration shows figure 9. The contact surfaces were described by sliding interfaces with a friction factor of 0,65. Variations of the factor showed minor influence on the calculated results of the compression test.

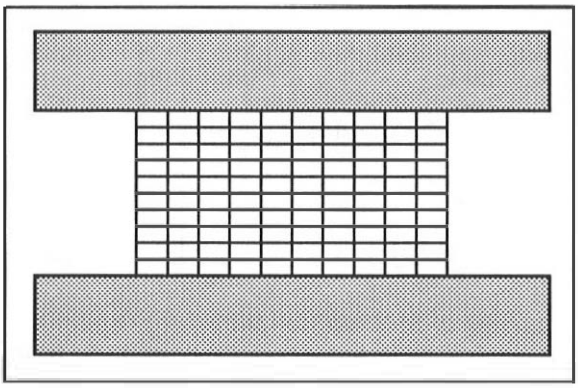

Figure 9. Finite element model of the skull bone compression test configuration

Figure 10 shows the contact force experimentally obtained and numerical calculated. The material model is able to simulate the force deflection curve from 0 to F1 (crush of diploë, fig. 10a) and from 0 to F3 (fig. 10b). The material type "non-linear fibre biphase solid" seems suitable for the simulation of anisotropic material properties of the human cranial bone. The material model was originally developed for composite materials.

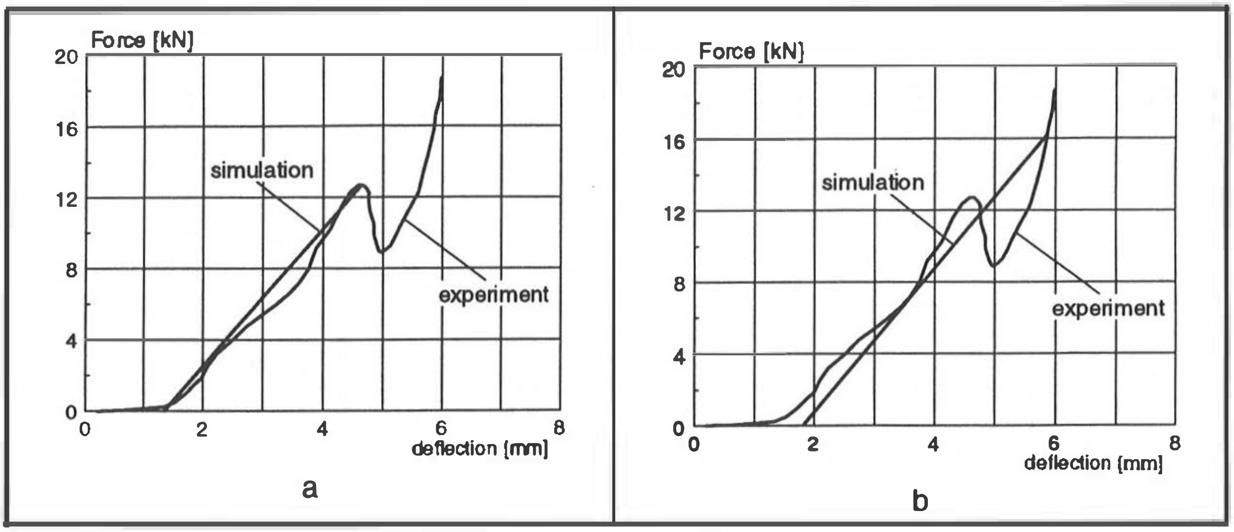

Figure 10. Comparison of experimental and calculated load deflection curves for bone in compression

Bending tests: The specimens were prepared as beams with a length of 50 mm, a width of 20 mm, a height between 3,5 and 9,8 mm and a radius ranging from 55 to 140 mm. Four representative tests with average results were chosen. Although the quantitative results varied, the force deflection curves indicate the same qualitative behaviour, with a linear slope until failure with a brittle-fracture. The bone specimens were modelled with 20 eight-node solid elements. The finite element model of the bending test configuration shows figure 11.

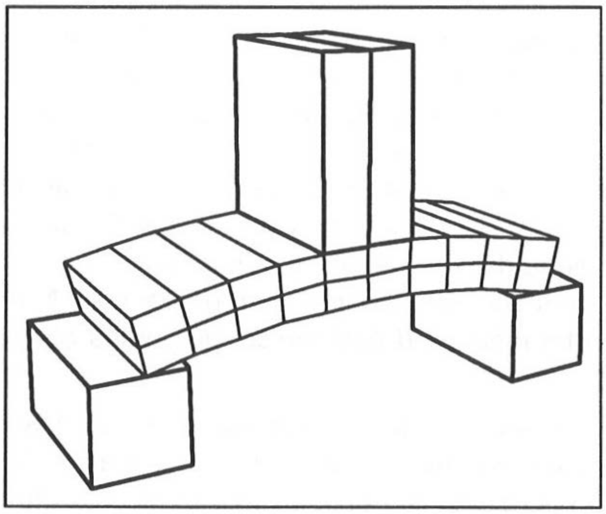

Figure 11. Finite element model of the three-point bending test configuration

Figure 12 shows for example the results of two different experiments and simulations. The difference between fracture force measured in the tests and calculated with PAM-CRASH is about 3%. The geometrical finite element model and the elastic-plastic material model simulate the linear phase of the bending test very sufficient.

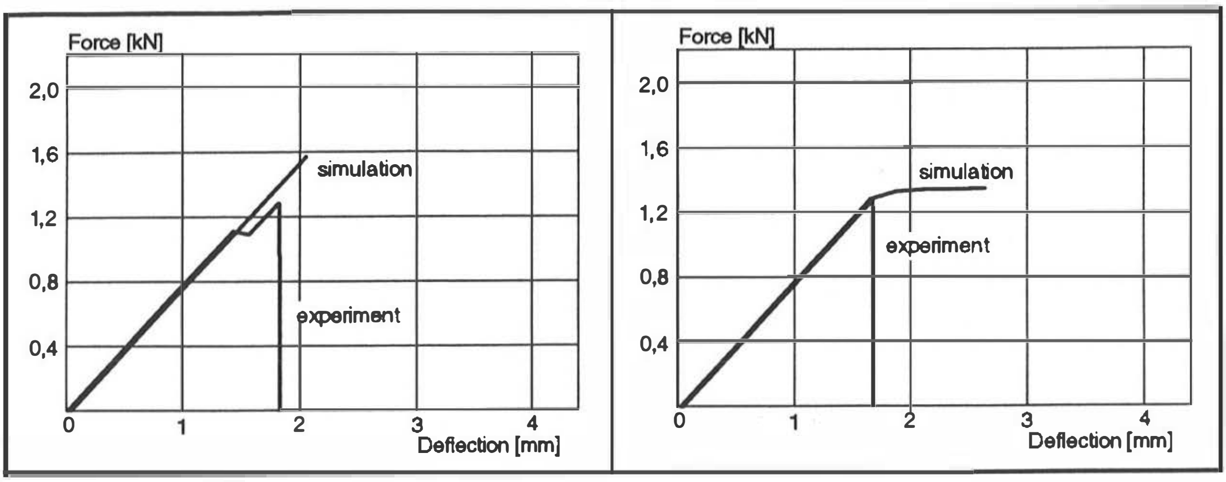

Figure 12. Comparison of experimental and calculated load deflection curves for bone in bending

SUMMARY

Head injury mechanisms are difficult to study experimentally due to the great amount of variables involved and the ethical problems using human cadavers and animals as surrogates of the living human. Finite element modelling is a comprehensive method to study the degree of human tolerance to head impact and to define injury probability thresholds.

To visualise the complex geometry of the anatomical structure of a human head two-dimensional Computer Tomography images were used. The CT scans were transformed into a three-dimensional finite element input file.

To integrate reliable material properties a series of tests was carried out in Heidelberg to determine basic mechanical properties of the bony structures of the human head using fresh bone specimens. The isolated human cranial bones were tested quasistatically in bending and compression. The results of these experiments were analysed, taking into account observed differences and varying structures. Statistical correlation of properties was made with geometrical, histological and physiological parameters. The finite element material model was validated by the same test methods. The results were compared with the experiments to determine reliable material properties. A numerical material model representing experimental measured mechanical properties of the human skull bone was developed.

Future extensions of the head model include the addition of an anatomical brain model based on Magnetic Resonance Images data. Once validated, the model will be used for detailed studies of biomechanics of head injuries related to impact and investigations of injury mechanisms that may replace and reduce laboratory experiments with human cadavers.

ACKNOWLEDGEMENTS

This research was supported in part by the Federal Highway Research Institute (BASt) under contract FP8921. The authors gratefully acknowledge the support from Engineering System International (ESI), Eschborn.

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