VALIDATION OF A FINITE-ELEMENT-MODEL OF THE HUMAN NECK

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ABSTRACT

This paper presents the development and the validation of a FE model of the human cervical spine with the explicite FE code PAM-CRASH™ as a collaboration between the Berlin University of Technology, Germany, and Engineering Systems International, Rungis, France. The model consists of deformable vertebral bones, intervertebral discs, articular cartilages and ligaments with an approximated anatomical structure. The geometry and the material model are described and the results of the simulations are compared with experimental results for frontal flexion, lateral flexion and compression of the human cervical spine in volunteer, cadaver and specimen tests. It is concluded that the presented FE model shows a good agreement with the test results. The simulation results underline the great influence of the elastic behavior of the vertebrae and the fixation boundary condition for the model on the global neck motion. The findings of this study show some possibilities of improving the biofidelity of the FE neck model.

THE HUMAN CERVICAL SPINE as the connecting part between the head and the torso combines two important tasks in its function. On the one hand it is responsible for the motion of the head and on the other hand the cervical spine protects the spinal chord from injuries. Therefore two aspects are of biomechanical interest:

- Motion analysis of the head-neck complex for a better evaluation of the head injury risk.
- Determination of the biomechanical limits of the cervical spine components to assess injuries of the components and the spinal chord

The structure of the human cervical spine and therefore its injury mechanisms are very complex. Studies with volunteers and cadavers have shown that an appropriate determination of injury mechanisms is rather impossible in experimental tests. Furthermore, a detailed and reproducible analysis of neck motion is necessary to improve the neck of anthropometric test dummies (ATD) for crash tests. Numerical simulations

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are a proper tool in addition to experimental tests to describe the biomechanical behavior of the human neck.

Different types of computational methods can be used for head-neck motion simulation. It seems that multi-body-systems (MBS) and the finite-element-method (FE) are most suitable for the representation of the complex mechanical behavior of the human cervical spine whereas two-pivot-models and continuum models can only represent the mechanical behavior in a most simplified way. In early studies two pivot models (three segments, two joints) and continuum models were used to describe the global head and neck motion (Terry and Roberts, 1968; Bosio and Bowman, 1986; Tien and Huston, 1987; Wismans et al., 1986; Wismans et al., 1987). To gain more confidence in the results of numerical simulation in the description of the head-neck motion during impacts several complex multi-body-systems were developed. Chen developed a 3D-model of the ligamentous human spine for static and dynamic simulation (1973). Another group of studies developing a MBS of the human cervical spine started with a two-dimensional model with all cervical vertebrae and the first three thoracic vertebrae (Reber and Goldsmith, 1979). The result of several modifications was a three-dimensional model (C1 - T2, including head and torso) with fifteen pairs of neck muscles (Deng and Goldsmith, 1987). The model of Deng and Goldsmith was implemented in the multibody code MADYMO by de Jager et al. (1994). MBS - models are suitable for the description of the global head-neck motion. In order to evaluate injuries of the human neck and to establish new biomechanical injury criterions for the neck it is necessary to describe injury mechanisms and to determine local mechanical parameters, such as stress, strain and pressure, as failure-limits to the different biomechanical components. The finite element method seems to be the most appropriate computational method of describing the global motion of the head-neck complex and of determing injury mechanisms for the neck. Kleinberger developed a FE-model of the human cervical spine (C1 - T1) with rigid vertebral bones, linear elastic material properties for the intervertebral discs, the articular cartilages and the ligaments.(1993). The simulation results were compared with results of frontal flexion and compression tests. Dauvilliers et al. presented a FE-model of the human cervical spine (C1 - T1) with rigid cervical vertebrae and linear elastic intervertebral discs (1994). Articular cartilages and ligaments were modeled as spring damper combinations, which was neccessary to obtain the proper kinematical behavior with regard to frontal and lateral flexion tests. Other FE models were developed to analyse the biomechanical behavior of parts of the human spine (Belytschko et al., 1974; Seenivasan et al., 1993; Tadano et al., 1993; Yoganandan et al., 1995).

This paper discusses the development of a FE model of the human cervical spine (C1 - C7), including all important vertebrae, discs, articular cartilages and ligaments. The FE model consists of deformable components to allow the integration of biological material properties and should be suitable to predict the motion for all relevant types of crash loads on car passengers. As a first part of the development the geometry model was created (Rade, 1993). The second part of the development led into an validated model of the three load cases frontal flexion, lateral flexion and compression (Nitsche, 1996).

DESCRIPTION OF THE HUMAN NECK MODEL

GEOMETRY MODEL - The FE model of the human neck consists of 3417 nodal points, 1852 solid elements and 86 membrane elements. The geometry of the model was generated on the basis of literature data (Rade, 1993). The following anatomical features are integrated into the model:

- Cervical vertebrae:
  Atlas, Axis, C3 - C7
- Intervertebral discs
- Articular cartilages
- Ligaments:
  Transversal ligament (LT)
  Anterior longitudinal ligament (ALL)
  Posterior longitudinal ligament (PLL)
  Ligamentum flavum (LF), including ligamentum nuchae
  Supraspinous ligaments (SSL)
  Intertransversal ligaments (ITL)

The cartilage-group and some ligaments are subdivided, taking into account, that the material properties might differ with regard to their position.

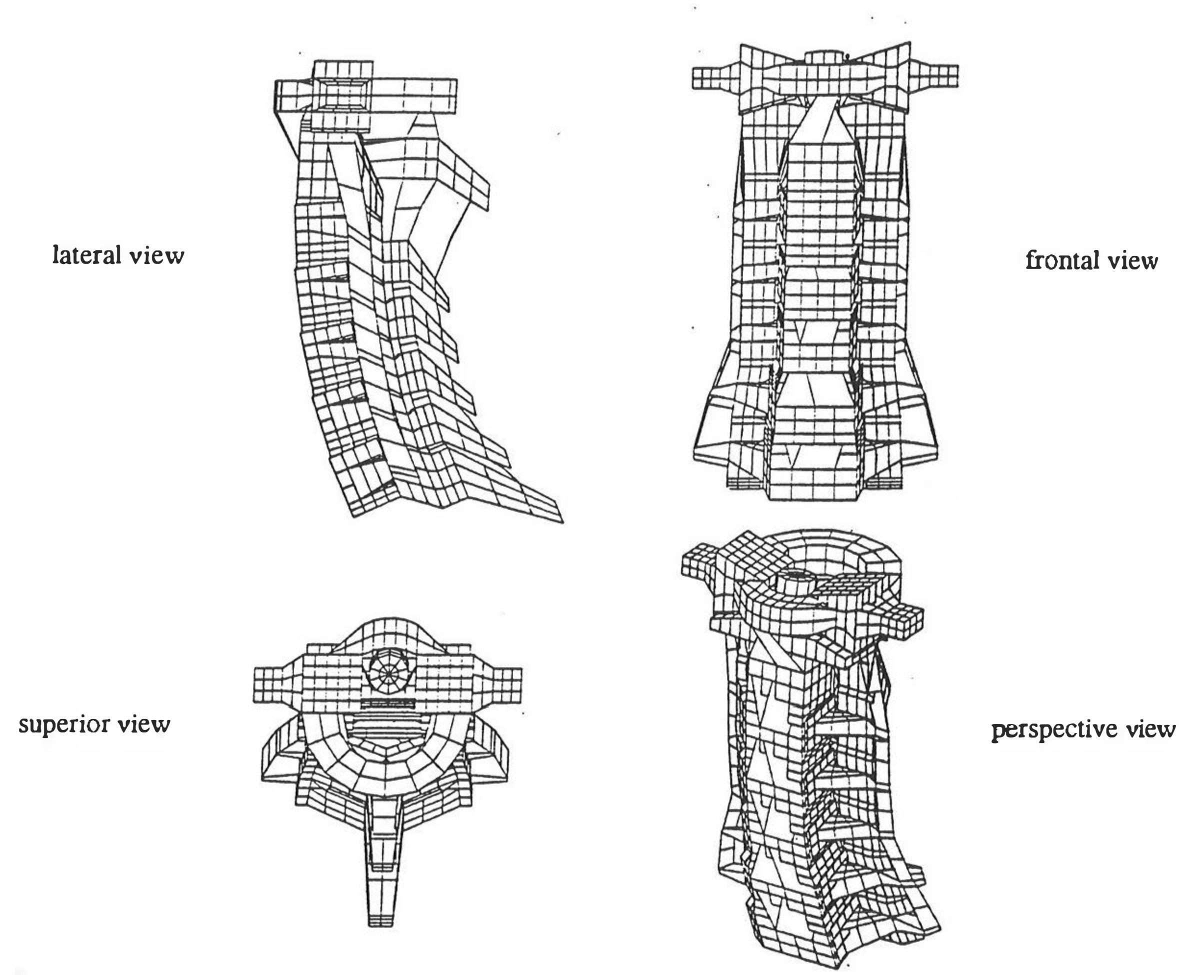
The curvature of the model represents the natural lordosis. The angle between C1 and C7 amounts to 25 degree. C1 and C2 are parallel, all other vertebrae are rotated by 5 degrees with regard to their upper vertebrae (Kleinberger, 1993, Nightingale et al., 1991). The intervertebral discs and the articular cartilages are all attached to the two adjacent vertebrae with one exception. The articular surfaces between C1 and C2 are only attached to the bony structure of C1 to allow a rotational motion between C1 and C2. Vertebral Bones, intervertebral discs and articular cartilages are modeled as solid elements. The ligaments are modeled as membrane elements. Different views of the model and its components are shown in Figure 1.

The muscles of the human cervical spine are not yet modeled because the greater part of the muscles are attached either to the head or to the thorax. Only the inner ring of cervical muscles are directly attached to the cervical vertebrae. However, the influence of the muscle contraction on the kinematic of the cervical spine can be neclegted, because the reaction time of a full contraction of the muscles is about 120 ms and therefore of no great influence on the neck motion (Burow, 1974). The maximum contraction forces are much smaller than the maximum crash loads on the human neck. Moreover, the influence of passive muscle action on the neck kinematic seems to be small compared with the effects of geometrical and material assumptions.

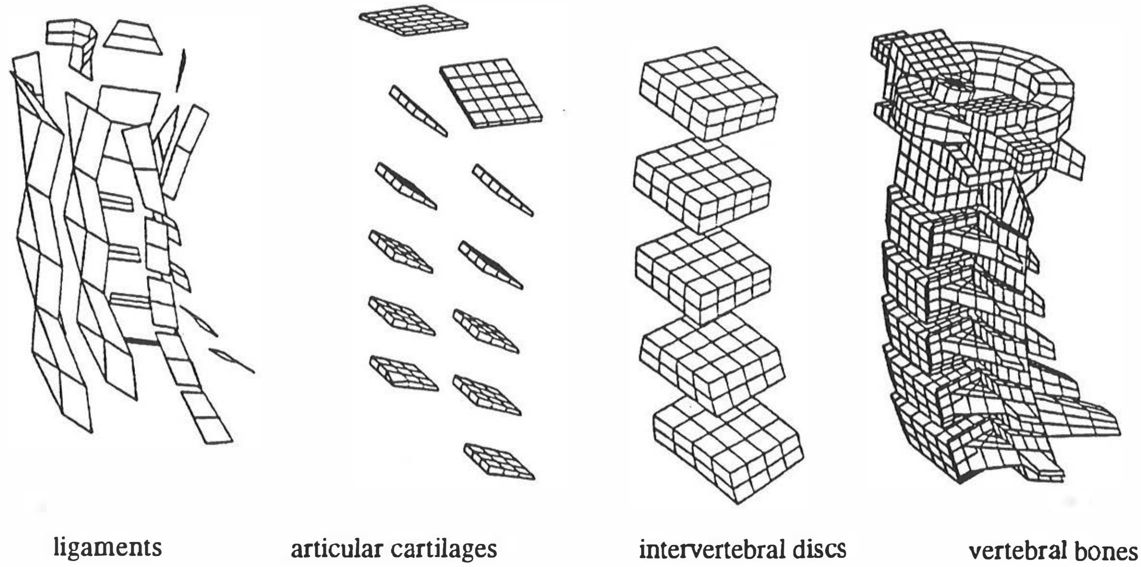
MATERIAL MODELS - There are two general problems concerning for the transfer of biological material properties into FE code input data. Biological material properties can generally be defined as nonlinear, viscoelastic, anisotropic and inhomogenous and are highly governed by other parameters like age, sex and constitution. It is often very difficult to apply forces to the separated biological specimen and to measure appropriate

Figure 1: Different views of the FE model and its components

complete neck model



neck model components - perspective view



values for all necessary load cases in order to determine all important mechanical parameters of biological materials. Therefore the biological material properties in the literature are often represented by stress-strain curves for only one load case or by global values like Young's modulus and failure load. On the other hand the necessary parameters for sophisticated FE-material models do not correspond to the parameters provided by extensive material tests with biological materials.

Many of the used material parameters are taken from the Yamada's review of properties of biological materials (1970). According to the above mentioned problems in data acquisition of biomechanical FE-materials the material properties of all components of the FE model are homogenous and linear elastic. The material properties of the vertebrae and the intervertebral discs are isotropic. The material behavior of the articular cartilages and the ligaments are anisotropic. According to the great differences in stiffness concerning compression and tension a contact is defined between the bony surfaces of the cartilage which creates an additional force when the cartilage is compressed in order to obtain a stiffer material behavior in compression than in tension. The ligaments are modeled with an anisotropic airbag material model. The fibres of the ligaments are in the direction of the applied tension force. Young's modulus was determined by a FE simulation of material test for human foot ligaments which were conducted at Wayne State University (Haug, 1995). A survey of the used material parameters is shown in Table 1. The ligaments which connect C1 and C2 are stiffer than all other ligaments because the mass of the head (5 kg) is applied to the center of gravity of C1. The articular cartilages between C1 and C2 are modeled with Young's modulus for compression because they can only transmit compressive forces.

Table 1: Material properties

	Young's Modulus [N/mm^2]	Poisson ratio
Vertebral Bone	300	0.25
Disk	100	0.4
Articular Cartilage	25	0.4
A.C. Atlas-Axis	500	0.4
Ligaments	2 * 52	0.22
Liga. Atlas-Axis	2 * 52	0.22

Experimental tests for both components showed remarkable differences for their material parameters, due to the differences of age, sex and constitution and to the different load cases. Furthermore, the conversion of nonlinear material properties into linear ones increases the field of possible material values. In order to obtain a validated kinematic, the disks are rather stiff in comparison with the values for the vertebral bones, but for both materials the values are within the biological range.

For the validation of the model two important effects of the assumptions made for the material properties must be considered. On the one hand the lacking of damping will limit the accuracy of prediction only to the first maximum of frontal and lateral bending. The rebound will be greater than in biomechanical tests. On the other hand, simulation results will not be dependent on the impact velocity because of the lacking of viscous effects.

VALIDATION OF THE HUMAN NECK MODEL

EXPERIMENTAL RESULTS - To cover many realistic load cases for the human neck in car crashes the model is validated against biomechanical experiments of frontal flexion, lateral flexion and compression of the neck. The tests were chosen in order to provide results which are comparable to the results of the simulation. The relative displacement in vertical and horizontal direction for frontal and lateral flexion and in vertical direction for compression seemed to be appropriate parameters for the validation of the model. Although the model could be validated against local parameters (stress, pressure, etc) there is no experiment that determines these local parameters. Therefore it is neecessary to validate the model against the global motion of the neck, and to assume that with similar results for experiment and simulation the local parameters stress and strain are also validated.

The model is validated against the volunteer test series of the Naval Biodynamics Laboratory (NBDL) (Wismans et al., 1986; Wismans et al., 1987) for frontal and lateral flexion and against compression experiments with separated human cervical spines (Pintar et al., 1989 a; Pintar et al., 1989 b; Nightingale et al., 1991; Pintar et al., 1995). For frontal and lateral flexion the relative displacement in horizontal and vertical direction was compared to that of the simulation. In the experiments the displacement of the occipital condyles (O.C.) and the center of gravity of the head (C.O.G. head) relative to a non-rotating T1 and the relative rotation angle of the head were measured. For the comparison of the experiment and the simulation it should be noticed that the T1-rotation is not taken into consideration and therefore should be taken into account for the interpretation of the simulation results. For the compression the compressive stiffness, the compressive failure displacement and the compressive failure load were compared with the results of the simulation. The test results are shown in Table2.

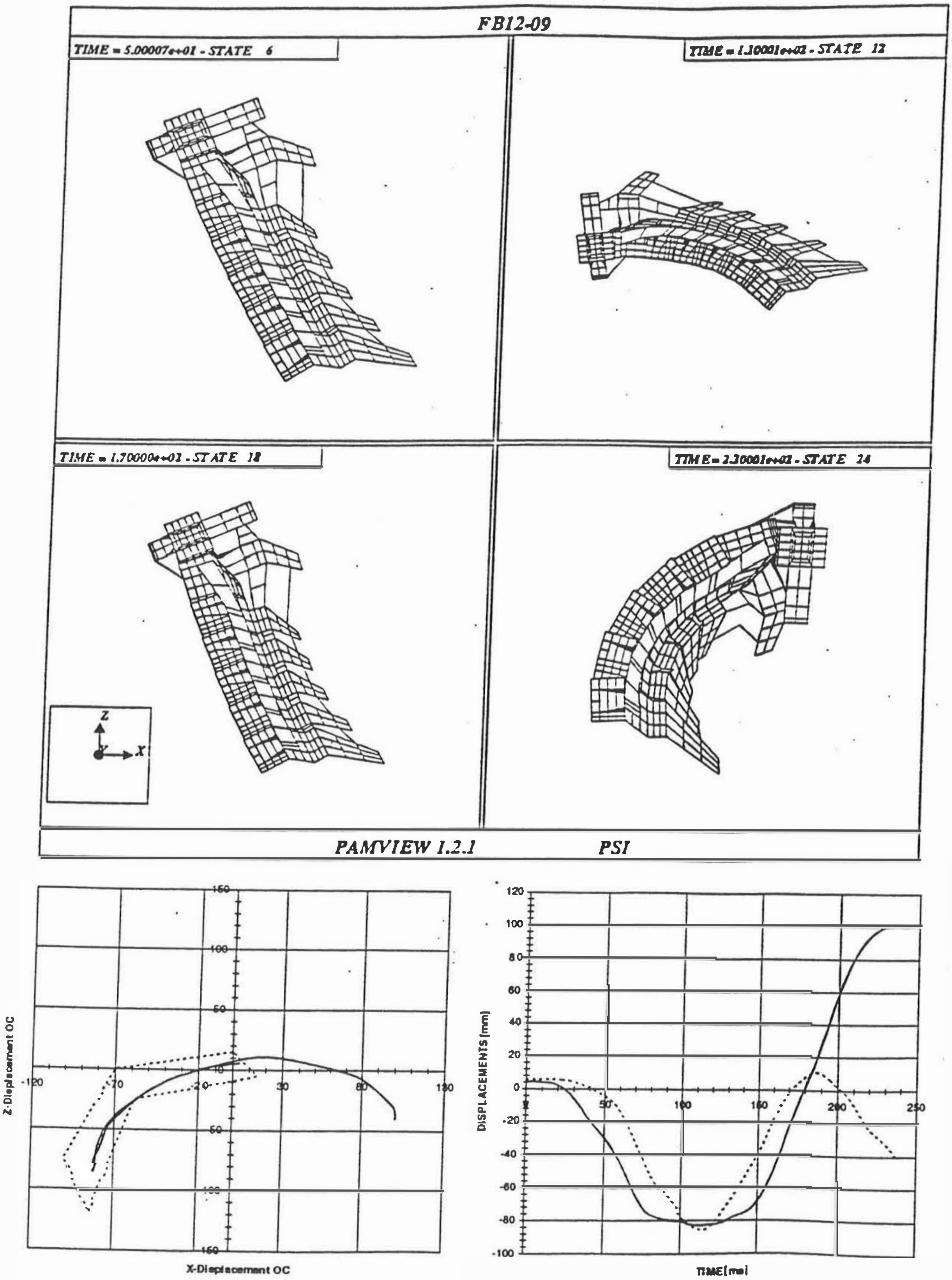
Table 2: Test results

frontal flexion: horizontal displacements O.C.	100 mm - 140 mm
frontal flexion: vertical displacements O.C.	150 mm - 215 mm
frontal flexion: horizontal displacements C.O.G. head	114 mm - 151 mm
frontal flexion: vertical displacements C.O.G. head	201 mm - 233 mm
frontal flexion: head rotation angle	68.4° - 94.9°
time for front. flexion max. relative to acceleration start	120 ms
lateral flexion: horizontal displacements O.C.	80 mm - 120mm
lateral flexion: vertical displacements O.C.	35 mm - 65 mm
compressive stiffness (mean values)	500 N/mm - 555 N/mm
compressive failure displacement (mean values)	14 mm - 18 mm
compressive failure load (mean values)	3326 N - 4810 N

SIMULATION RESULTS - For the frontal and lateral flexion simulation the lowest vertebral bone C7 was rigidly fixed. The mass of the head (5 kg) was added to C1 in an approximated position of the center of gravity of the head. The sled acceleration profile was imposed to the C.O.G. of the head with a maximum acceleration of 15 g for frontal flexion and 7 g for lateral flexion (Wismans et al., 1986). Three different series of simulation were performed for frontal flexion to evaluate the influence of vertebral bone elasticity and the influence of T1-rotation. A linear elastic joint was added as a boundary condition for C7, which allowed a maximum rotation about the y-axis of 23 degrees. Table 3 shows the simulation results of frontal flexion in comparison with the experimental results. Figure 2 shows different steps of neck motion, the trajectories and XZ-displacements over time of C1 for frontal flexion with elastic vertebrae and elastic joint for C7 rotation. The dotted line in the trajectory graph indicates the field of results of volunteer tests (Thunnissen et al. 1995).

As to for lateral flexion two different sets of simulations were performed to evaluate the influence of elastic vertebrae. An elastic joint for lateral rotation of C7 could not be integrated because of lacking information about the rotation angles in the tests. Table 4 shows the simulation results of lateral flexion in comparison with the experimental

Figure 2: Visualization of the motion of the model and trajectories (left) and XZ-displacements over. time of C1(right) for frontal flexion with elastic vertebrae and elastic joint for C7 rotation



results. Figure 3 shows different steps of neck motion, the trajectory for C1 and the rotational angles of C1 over time for lateral flexion with elastic vertebrae.

Table 3: Simulation results of frontal flexion

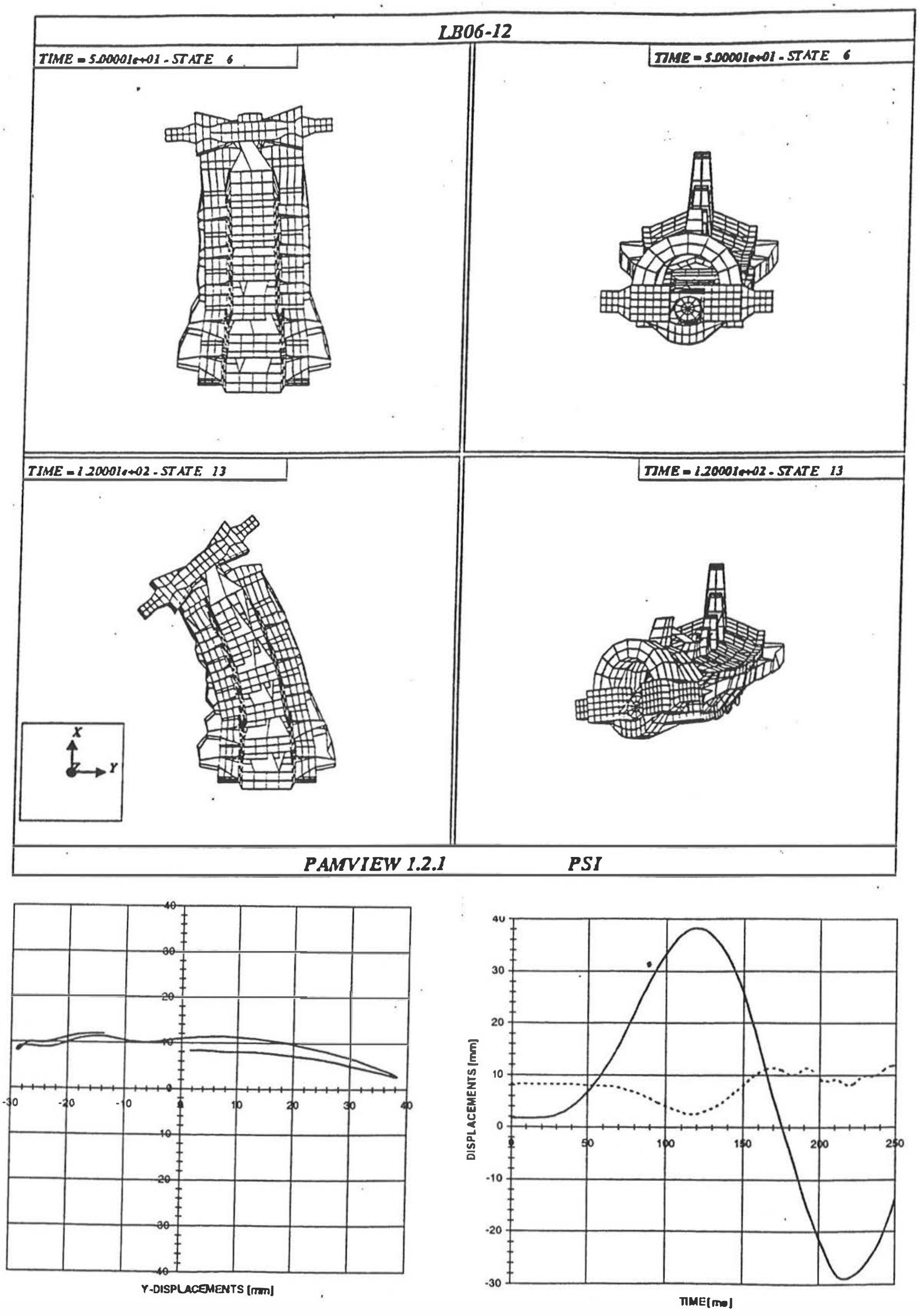
	Time	Maximum flexion of C1		Maximum Flexion of C.O.G.-Head		Maximum C7-rotation
	maximum flexion [ms]	forward [mm]	down-ward [mm]	forward [mm]	down-ward [mm]	[degree]
Test	120	100-140	150-215	114-151	201-232	68.4-94.9
Rigid Bones	85	50	21	92	43	48
Elastic Bones	100	74	61	130	120	94
Elastic Bones, C7 Rotation	110	85	91	140	163	100

Table 4: Simulation results of lateral flexion

	Maximum forward flexion of C1 [mm]	Maximum downward flexion of C1 [mm]
Test	80 - 120	35 - 50
Rigid Bones	14	1
Elastic Bones	36.6	6

For compression three different sets of simulation were performed to evaluate the influence of the initial position in compression testing. The different initial positions were: (a) natural lordosis with C7 in a perendicular position according to the load direction; (b) without natural lordosis, all vertebrae prependicular to the load direction; (c) natural lordosis with C1 perpendicular to the load direction. In all cases C7 was fixed and the C.O.G. of C1 was displaced vertically with a constant velocity. Compression

Figure 3: Visualization of the motion of the model, trajectory of C1 (left) and angles of C1 over time (right) for lateral flexion with elastic vertebrae



force and displacement were measured to determine compressive stiffness, compressive failure load and compressive failure displacement. Table 5 shows the results of the simulation for all cases in comparison with the results obtained in the experiments. Figure 4 shows final deformations and force-deformation-curves of all three initial positions.

Table 5: Simulation results of compression

	curved, C7 vertical	parallel	curved, C1 vertical	experimental results
Stiffness[N/mm]	500	1000	700	500 - 555 N/mm
Failure Load [N]	3000	6500	4500	3326 N - 4810 N
Failure Deformation [mm]	8	8	8	14 mm - 18 mm

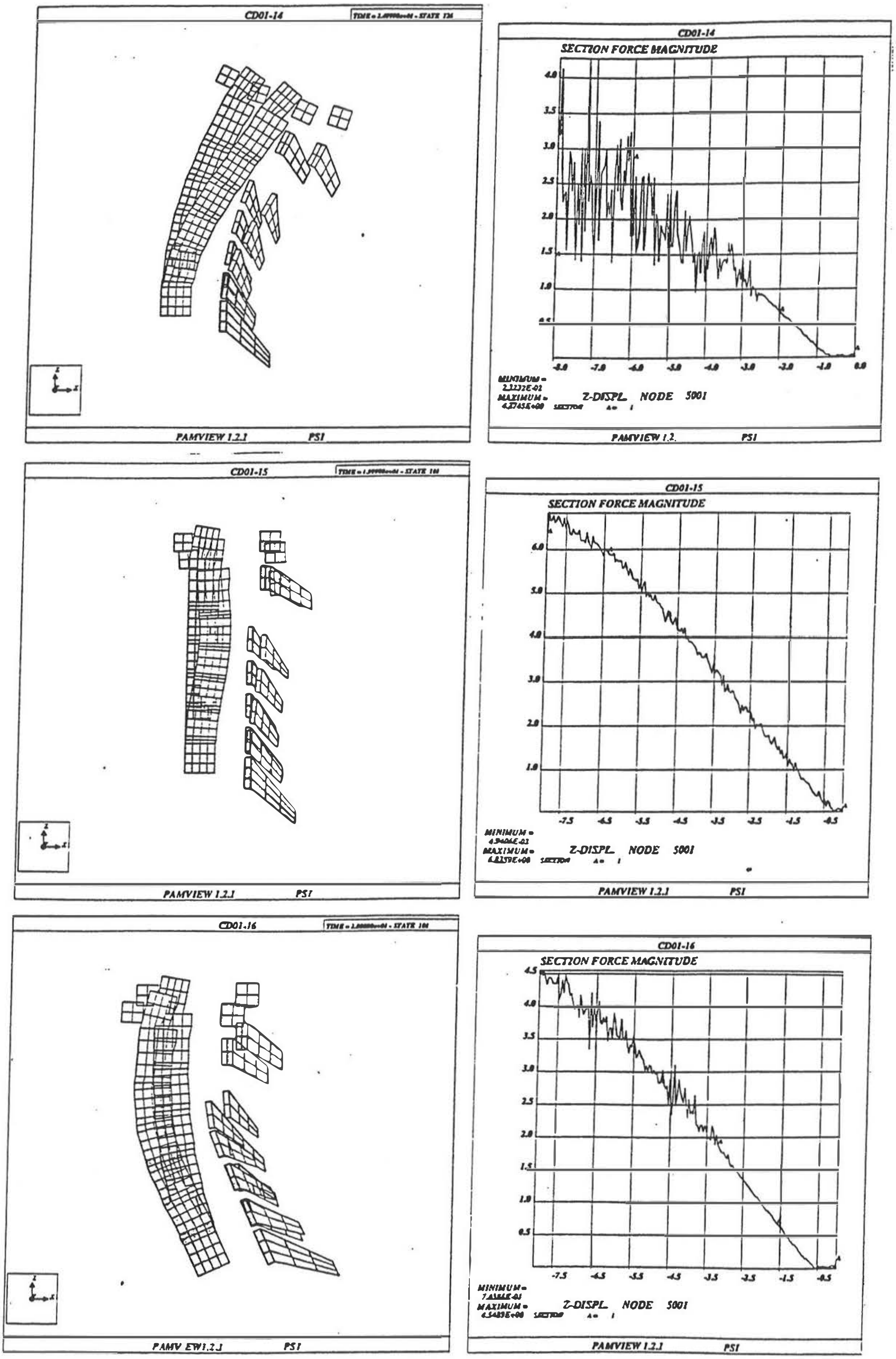
DISCUSSION

The results of the simulations with the human neck model quite agree with the global neck motion in the tests frontal flexion, lateral flexion and compression and also with the maximum displacements in frontal flexion and compressive stiffness in compression. However, the maximum displacements in lateral flexion are less in accordance with the test results.

The comparison of the results of simulation with and without elastic vertebrae shows for all load cases that the influence of elastic material properties on the vertebral bones is not necligible for proper reproduction of the biomechanical behavior of the human neck. Furthermore, the rotation of C7 influences the global motion in a very strong way, so that an appropriate boundary condition for the fixation of C7 is absolutely necessary to achieve good agreement with the test results. The displacements of Cl and C.O.G. of the head show, in comparison with test results, that a rigid connection between C l and head is sufficient for the validation of the global motion of a the FE-neck model.

The simulation results of the frontal flexion show a good correspondence with the test results both for quality and quantity. The motion is a combined tension and compression of all deformable elements of the model. The neck model seems to behave little stiffer than a human neck in volunteer tests, which can be explained by the different methods of displacement measurement in the simulation and in the tests. The inital distance between the first thoracic vertebra as the pivot point of the rotation and the occipital condyles as measuring points is larger than the distance between the pivot point (C7) and the measuring point (Cl) in the simulation. Therefore the possible displacement as a

Figure 4: Left side: Visualization of the initial state (white) final deformation (gray) of the neck model; Right side: Force-displacement-curve for compression; Upper row: Curved, C7 vertical; Middle row: Parallel; Lower row: Curved, Cl vertical



consequence of rigid body rotation around the pivot point is larger in the tests than in the simulation. In addition to that, C1 and C7 are modeled as rigid bodies in order to achieve a proper force distribution in the remaining part of the neck model. The deformation of C7 and the intervertebral discs between C7 and T1 would increase the global displacement of the neck.

The simulation results of the lateral flexion show less good quantitative agreement with the test results, which can partly be explained by the lacking information about the T1 rotation in the tests. The qualitative analysis shows that the model is suitable for representing the complex motion of the human neck for lateral flexion. The motion of the model shows that the lateral flexion is a combination of local bending, compression, tension and torsion between all deformable elements of the model. Again, the lacking of detailed information about local displacements and rotations defeats at this time a proper quantitive assessment of the model's behavior for lateral flexion.

The simulation results of the compression show good qualitative and quantitative agreement with the test results. According to findings in the experimental studies the straight model without natural lordosis behaves stiffer than the models with natural lordosis. In all cases the model deforms according to the experimental prediction. The largest deformation appears in the region of the third cervical spine (C3). The maximum failure displacement in the simulation is lower than in the tests, which can be explained by the use of linear elastic material properties and the stop of the calculation when a negative brick volume is reached.

OUTLOOK

As a next step, the existing model will be modified in order to achieve an increased biofidelity. On the basis of the findings of the current study local geometries will be refined according to their influence on the global motion of the neck. Furthermore, other material models which are nearer to the mechanical behavior of biological components will be integrated into the model. For a reliable quantitative validation of the model, especially for lateral flexion cases, it is necessary to analyse the existing experiments in detail in order to obtain new information about local displacements and rotations. This could be done in a co-operation between the testing and the simulating institutions.

The final objective is to link the validated FE neck model with a validated FE head and FE thorax model. For the future, there will be four possibilities of using a complete human head-neck-thorax-model:

- Greater reliability in car safety assessments by an increased biofidelity of the human head-neck-junction.
- Increased biofidelity of crash dummies by a better understanding of the head, neck and thorax kinematics provided by a head-neck-thorax-model.
- Development of new injury criteria for the head and the neck.
- Direct injury assessment for the head, the neck and the thorax on the basis of local mechanical data (e.g. stress, strain, pressure) in realistic car crash simulations.

CONCLUSIONS

This study illustrates that the presented FE model is generally suitable to describe the complex motion behavior of the human cervical spine for different load cases which can occur in car accidents. The differences between the results obtained in the experimental tests and in the simulations can be explained by the various assumptions and simplifications which were made at an early stage of the validation of the model.

The simulations showed the great influence of the elasticity of the vertebral bones on the global motion of the neck model and the need for a proper representation of the fixation boundary condition for C7. A rigid connection between C1 and the head seems to be sufficient for the validation of the neck model.

As to frontal flexion, the model shows good qualitative and quantitative agreement with the test results. The stiffer behavior of the model can be explained by different local coordinate systems for the tests and the simulations and the lacking of deformation for the rigidly modeled vertebrae C1 and C7.

As for lateral flexion the model shows the complexity of the motion and a less good quantitative agreemeent with the test results. Further quantitative assessment is still defeated by the lacking information about proper boundary conditions for the fixation of C7.

With regard to compression, the model shows good qualitative and quantitative agreement with the test results. The deformation patterns of the simulations correspond to experimental findings.

The development process pointed out two major problems in biomechanical simulation. The conversion of biological material properties into FE material model parameters is, at present, very difficult. As a consequence of the need for biomechanic material models it is important to improve the interface between the parameters obtained in experiments with biological materials and the parameters needed for FE material models. The validation of a separated FE model of the neck leads into well defined but, according to experiments with volunteers and cadavers, artificial and not realistic boundary conditions. Therefore a proper measuring equipment and a detailed analysis of experimental results is necessary to determine, for a validation, appropriate boundary conditions of the model. Sometimes the provided information in the literature is not sufficient.

The findings of this present study also showed aspects for a possible improvement of the biofidelity of the model. At the next stage local geometries and material models will be changed in order to increase the reliability of the model and the model can be assembled with FE models of the head and the thorax to represent a biomechanical upper body motion.

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