A HUMAN MODEL FOR LOW-SEVERITY REAR-IMPACTS

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ABSTRACT

Neck injuries resulting from rear-end collisions rank among the top safety problems and have serious implications for society. In an attempt to minimize the severity of neck injuries in such accidents, an increasing number of studies to evaluate the effectiveness of head restraints has been performed. In these studies, volunteers, crash test dummies, and mathematical dummy models were used. In addition, a limited number of mathematical models of the human body was used. However, to the best of our knowledge, these models were not validated in an environment comparable with a rear-end collision.

The objective of this study is to develop a mathematical model of a seated occupant and to better understand the biomechanical response of the spine and the occupant's interaction with the seat during rear-end collisions. For this purpose, a 3D mathematical model of a 50^{th} percentile sitting adult male is developed for use in simulations of rear impacts. Special attention is paid to the modelling of the spine, including the neck, and the occupant's interaction with the seat. To obtain insight into its biofidelity, the model's response is compared with rear-end sled tests with volunteers and human cadavers at a ΔV of up to 30 km/hr. The model is then used to study and quantify the motion of the spine in low and medium severity rear-end collisions. This study revealed that, during the "torso loading phase", the pelvis was lifted from the seat while the vertical motion of the T1 vertebral body relative to the vehicle was slight. Spinal compression occurred during this phase, but it remained slight. Although a thorough validation of the model developed was not possible due to lack of experimental data, it can be concluded that this model has the potential to become a powerful tool for parametric studies to aid in a seat design process.

CURRENT KNOWLEDGE of the injury mechanism of whiplash is limited. Computer models of the human body can be useful to study whiplash. Several mathematical models of the neck have been developed, and some possible injury mechanisms have been proposed (2). However, what is lacking in the current literature is accurate information on the displacements and accelerations of the first thoracic vertebral body (T1) during rear-end impacts. This is important since the T1 kinematics can be considered as "input" to dynamic models of the neck. From an engineering point of view, it is reasonable to assume that the T1 kinematics is a result of the following two mechanisms. First: rigid body motion of the entire spine, including the pelvis; and second: d-eformation of the spine itself. Accurate data on T1 displacements are difficult to obtain from experiments because of possible interference of the instrumentation with the seatback. If volunteers were (at low values for ΔV), an additional complication is the fixation of the instrumentation to T1. In the current study, a mathematical model of the lumbar and thoracic spine is developed and added to an existing rigid body model of the neck and head.

OTHER WORK - Several mathematical models of the human spine already exist. For example, Ome and Liu (15) developed a 2-D model of the human spine which incorporated the axial, shear and bending resistance of the discs, and modeled each vertebrae as a rigid body with three degrees of freedom. This model was later adapted by Prasad and King (16) and validated by comparison with cadaver drop tower tests (loaded in a vertical direction). Belytschko and Privitzer (1) developed a 3-D model in which the human body is represented by a collection of rigid bodies interconnected by deformable elements. This model was validated by comparing its vertical impedance with experimental measurements by Vogt *et al.* (22). None of these models were validated for a (horizontal) loading situation similar to a rear-end or frontal impact.

The model that probably comes closest to the one presented in this paper is the spine model developed at Chalmer's University (9), in which the human spine is represented by 24 rigid bodies, interconnected by hinges. The geometry of the spine elements was obtained from Robbins (17). The spine model was implemented in the 50th percentile sitting male Hybrid III dummy database (21). Non linear rotational stiffnesses and damping coefficients were defined for each joint. Although this study represents a major step forward in the study of head/neck responses during low severity rear-end collisions, it still has some limitations. For example, the arms and shoulders are not modelled as separate bodies, i.e. the mass of the arms and shoulder was added to the mass of the spine. Also, the validation of this model is mainly qualitative. This was done by applying a rear-end collision pulse to the model, similar to the pulse used by McConnell *et al.* (11), (ΔV = 8 km/h), and comparing the resulting head angles. However, as the authors indicated, the curvature of the upper part of the spine appears to differ between the model and the volunteer in McConnell's study. This may affect the validity of the interaction between the seat-back and the back of the occupant model.

OBJECTIVES - The objectives of this study are as follows. Firstly, to develop a mathematical model of a seated car occupant, including the seat. Secondly, to obtain insight into the model's behaviour compared with the response of human volunteers and human cadavers in rear-end sled test experiments. Thirdly, to use the model developed to get an impression of the occupant's interaction with the seat during low and medium severity rear impacts. More specifically, to get an idea of the displacements of the T1 vertebral body and the pelvis; the elongation of the spine; and the changes of the shape of the spine will be investigated.

METHODS

In the following, the human model and the seat model is described. Next, although experimental data on rear-end sled tests with volunteers or post-mortem human subjects (PMHSs) are sparse, the model predictions are compared with the results of experiments to obtain insight into the biofidelity of the model. Finally, six simulations are performed with the human model supported with the seat developed, at different levels of severity.

HUMAN MODEL - The human model consists of 3 components: a thoracic+lumbar model representing the human spine up to T2; a head-neck model (starting with T1); and finally, the remaining body parts (such as arms and legs). Starting point was a validated MADYMO model of a 50th percentile sitting adult male Hybrid III dummy (21). In this model, the bodies and joints representing the entire spine and the head were replaced by a more detailed representation as described below (Fig. 1).

Lumbar+thoracic spine model - The spine model consists of 16 bodies representing the lumbar and thoracic vertebrae (not including T1). The intervertebral discs are incorporated in these bodies. The location and the orientation of the spine joints in the mid-sagittal plane are chosen according to the anthropometry of a 50th percentile seated adult male (17) (Table 1).

Table 1a - Size and orientation of vertebral bodies of lumbar+thoracic spine model (inferior portion). Note that the orientation of the joints is given relative to the inferior body.

	L5	L4	L3	L2	L1	T12	T11	T10
Height (cm)	3.3 5	3.20	2.90	3.15	2.95	3.30	3.20	3.20
Orienta-tion (deg)	5.0*	-11.5	-11.5	4.5	4.5	6.2	6.3	5.4

* w.r.t. pelvis body.

Table 1b - Size and orientation of vertebral bodies of lumbar+thoracic spine model (superior portion). Note that the orientation of the joints is given relative to the inferior body.

	T9	T8	T7	T6	T5	T4	T3	T2
Height (cm)	3.2 0	3.00	2.45	2.90	2.75	2.15	1.85	1.90
Orienta-tion (deg)	4.7	4.3	6.3	5.2	5.1	5.0	4.8	4.7

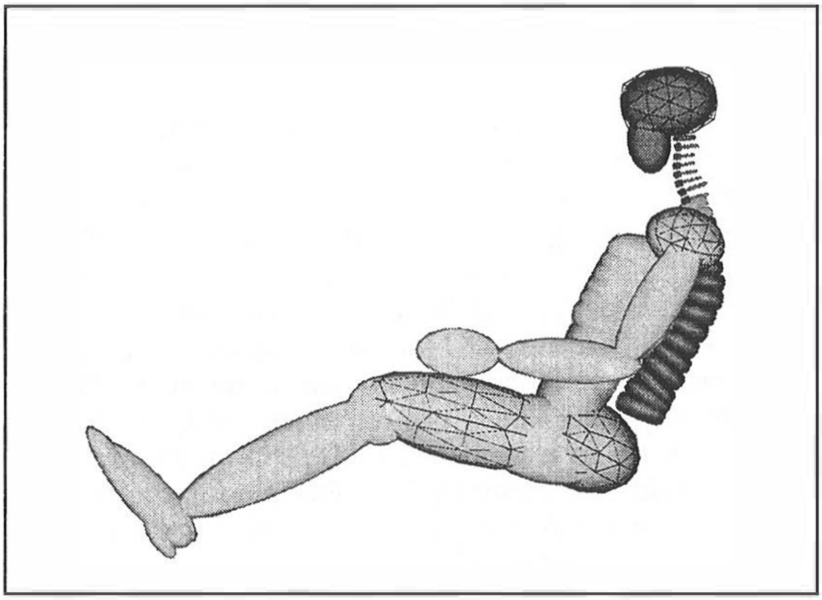


Fig. 1 - Human Model

The total mass of the lumbar+thoracic spine model is equal to 12.6 kg (based on the total mass of the lumbar spine and the spine box of a Hybrid III dummy), and is equally divided among the 16 bodies, resulting in 0.79 kg per vertebra. Mass moments of inertia are set at 0.001 kgm^2 (1). Although the spine model is in principle a three-dimensional model, only two degrees of freedom are allowed in this model: flexion/extension and axial elongation/compression. As this study focuses on motions in the sagittal plane, the other four degrees of freedom (lateral bending; axial torsion; anterior/posterior shear; and lateral shear) are suppressed. Regarding the flexion/extension motion, linear stiffnesses are defined for each of the joints to represent the soft tissue and muscles acting at the joints (Table 2). The stiffness increases non-linearly close to the range-of-motion (ROM) values. Rate-dependent behaviour is represented by constant damping coefficients (Table 2). With respect to the axial properties, linear stiffness and damping is assumed (Table 2). All stiffness and damping parameter values are obtained from Prasad and King (16). The range of motion of the thoracic+lumbar spine model is chosen at 105 and 60 degrees, for flexion and extension, respectively, according to Kapandji (10). These ROM values are equally divided between the lumbar and thoracic spine (5), and equally distributed among the vertebrae in these regions.

Head-neck model - The so-called "global model" developed by De Jager (8) is used to represent the head and cervical spine. This model comprises nine rigid bodies for the head, the seven cervical vertebrae and the first thoracic vertebra (T1). The bodies are connected by three-dimensional non-linear visco-elastic elements with load-displacement characteristics derived from recent experimental data on cervical motion segment behaviour (8). For simplicity, muscle behaviour was lumped into the intervertebral joint stiffnesses.

Table 2a - Joint properties for lumbar+thoracic spine model (inferior region)

	L5/S1	L4/L5	L3/L4	L2/L3	L1/L2	T12/L1	T11/T12	T10/T11
$K_{fl\text{-}ext}$ Nm/rad	678	678	678	678	678	678	678	1356
K_{axial} kN/m	1429	1429	1429	1429	1429	1429	1429	1429
$B_{fl/ext}$ Nm/rad/s	1.13	1.13	1.13	1.13	1.13	1.13	1.13	2.26
B_{axial} N/m/s	1790	1790	1790	1790	1790	1790	1790	2680

Table 2b - Joint properties for lumbar+thoracic spine model (superior region)

	T9/T10	T8/T9	T7/T8	T6/T7	T5/T6	T4/T5	T3/T4	T2/T3	T1/T2
$K_{fl\text{-}ext}$ Nm/rad	1356	1356	1356	1356	1356	1356	1356	1356	1356
K_{axial} kN/m	1841	1841	1841	1841	2958	2958	2958	2958	2958
$B_{fl/ext}$ Nm/rad/s	2.26	2.26	2.26	2.26	2.26	2.26	2.26	2.26	2.26
B_{axial} N/m/s	2680	2680	3580	3580	4470	4470	4470	4470	4470

This model was validated by comparison with the response of human volunteers to frontal impacts. Although linear and angular accelerations correspond satisfactorily, the total head rotation was too great. Therefore, the stiffness characteristics for flexion/extension are changed slightly in order to meet the required maximum head rotations (Fig. 2).

SEAT MODEL - The seat model is made up of a steel frame with deformable cushions. It consists of three bodies: a seat, a seat-back, and a head restraint. (Fig. 3). The following parameters can be varied: seat-back angle; head restraint position (horizontal and vertical); rotational stiffness of the seat-back joint; and the stiffnesses of the seat cushion (lower, middle, upper), the seat-back and the head restraint. In the current study, the joint between the head restraint and the seat-back is locked.

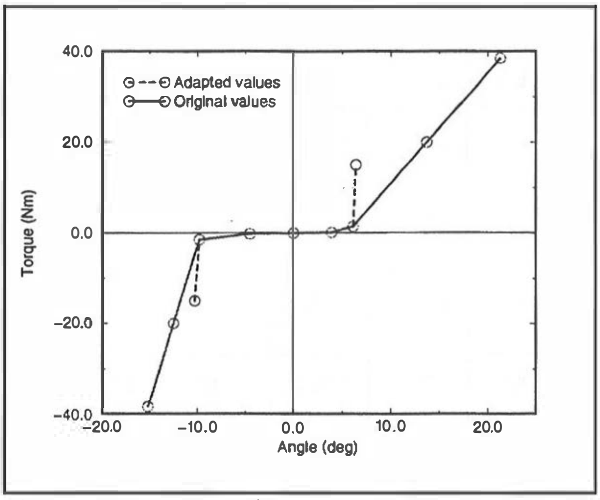


Fig. 2 - Adaptation of joint characteristics in neck model

Interaction with seat - It is expected that the deformation of the seat has a considerable influence on the response of the human model. Generally, the seat would be modelled using finite elements (FEM). However, this requires large calculation times, which makes such a model less suitable for parametric or optimization studies. Therefore, an alternative method is employed, in which arbitrarily shaped objects are defined by covering the surfaces with triangular shaped facets. This has the advantages of the finite element method as the geometry can be modelled with any desired level of accuracy, and distributed (rather than point) contact loads are calculated. However, no additional degrees-of-freedom are introduced into the model, so that the method requires only a fraction of the computational time of a finite element model. In the human model, the lower torso, femurs, shoulders, skull and back are modelled using these arbitrarily shaped surfaces. The seat cushion, seat-back cushion, and head restraint are also modelled in this way.

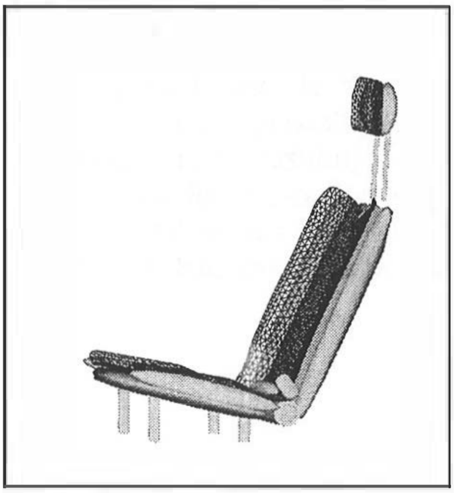


Fig. 3 - Seat Model

TEST RUNS - To obtain insight into the dynamic behaviour of the complete model in a rear impact situation, its performance was compared with several rear-end sled experiments at low and high velocities with volunteers and post-mortem human subjects sitting in a seat with a rigid back support (7). After considering the overall kinematics of the model, the head angle and the vertical displacement of T1 are considered. Also, the static performance of the spine was investigated. The experiments were simulated by placing the human model on a rigid seat, and applying the appropriate pulses. Three different pulses were used with values for sled-ΔV ranging from 6 to 30 km/hr.

Overall dynamics - Figure 4 illustrates the overall dynamic behaviour of the human model during a rear-end impact with a ΔV of 20.4 km/hr without a head restraint (a_{ave} = 35 m/s^2). The time interval between the images is 60 ms. Several phenomena that were also observed during experiments can be noted. For example, head extension is preceded by rearward translation of the head (often called "head lag", (20)). Also, the pelvis is lifted from the seat when the head angle is close to its maximum value. Although not very clear from the figure (due to obstruction of the arms), the spine is first straightened (up to about 180 ms, 3rd image) and then flexes. Similar results were obtained when the higher and lower severity pulses were used.

Static performance - The static performance of the spine (S1-T1) in flexion and extension is tested by comparison with experimental data from volunteers, which were the basis of a response corridor defined by Hoofman et al (7). Considering the rotational degree of freedom only, it can be seen that the model fits well within the corridor for both flexion and extension (Fig. 5).

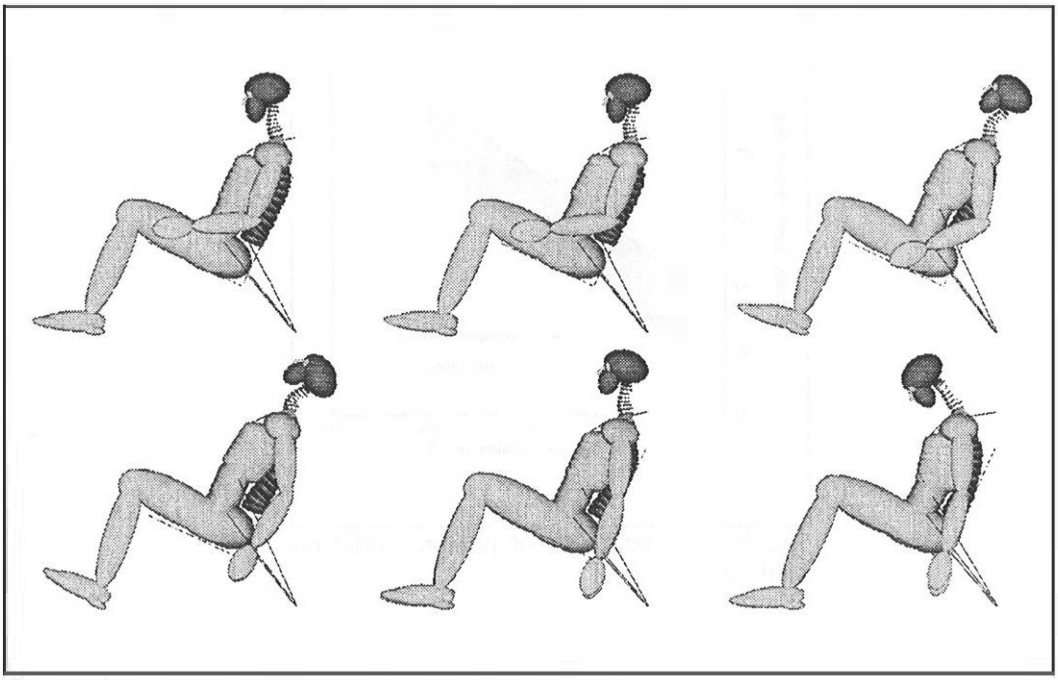


Fig. 4 - Overall kinematics of human model on rigid seat

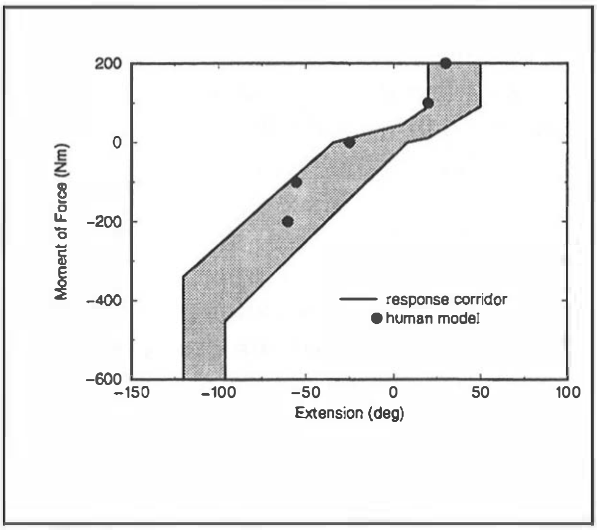


Fig. 5 - Static performance of spine model

Head angle - Based on various experiments with volunteers and cadavers, Thunnissen defined a linear response corridor relating the maximum head rotation to the average acceleration of the sled (19, 20). Note that the head angles are defined relative to the torso and consequently include the T1 vertebral body rotation. As can be seen in Figure 6, the maximum head rotations for the three test runs are within the defined corridor, representing a fully relaxed volunteer.

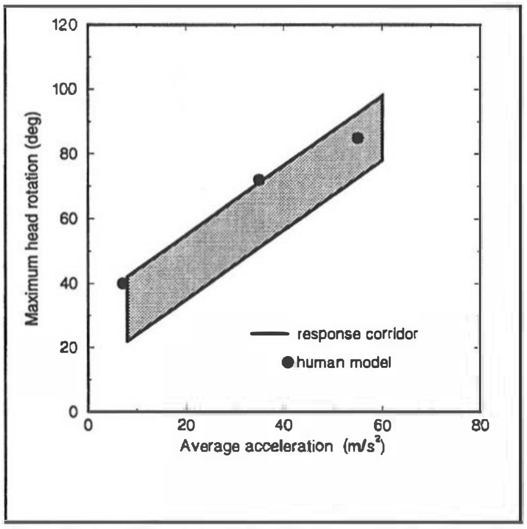


Fig. 6 - Head angle of human model on rigid seat

Vertical T1 displacements - Another source of comparison is the vertical displacement of the T1 vertebral body. Hoofman *et al.* (7) defined a corridor for this parameter based on two low-severity (ΔV = 8 km/hr) rear-end studies with volunteers. The corridor was defined between 10 and 30 mm of vertical displacement along the seat-back axis. The vertical displacements in the three simulations are 1 mm, 13 mm and 25 mm (expressed along the vertical axis of vehicle coordinate system), and therefore correspond reasonably well with the experimental situation. Note that these numbers will be somewhat larger if expressed along an axis attached to the (rotating) seat-back.

SIMULATIONS - For the simulations with the human model placed in the developed (non-rigid) seat, three different pulses are used, with and without a head restraint, resulting in a total of six cases (Table 3). The occupants are secured by a three-points belt. Simulations are run for 200 msec, except for the low severity cases, which are run for 300 msec. Regarding the seat model: the rotational joint stiffness of the seat-back is 1250 Nm/deg (18). The stiffness of seat-back and seat cushion is set at 27 kN/m up to about 1.5 cm of indentation, after which the stiffness increases to a value of 704 kN/m (3). The initial angle of the seat-back is 25 degrees to the vertical. The coefficient of friction at all contacts between the human model and the seat is set at 0.29. The head restraint is placed at 7.5 cm from the back of the head. Its height is adjusted so that the top surface of the head and of the restraint are aligned.
The following output parameters are obtained:
- joint rotations in the spine
- axial displacements in the spinal joints
- displacement of the upper thoracic vertebra (T1)
- displacement of the pelvis
- head rotation
- contact forces with the head restraint
- shear forces at the joint between the upper cervical vertebra and the head (OC joint).

Table 3 - Test conditions for simulations

run name	a_{ave} m/s^2	Δv km/hr	head-restraint
nhr_l	7.1	5.7	no
nhr_m	35	20.4	no
nhr_h	55	30.3	no
hr_l	7.1	5.7	yes
hr_m	35	20.4	yes
hr_h	55	30.3	yes

RESULTS

SPINE DEFORMATION - The first five output parameters, as defined in the Methods section, are presented below (see also Table 4). First, note that for the low severity runs, no contact with the head restraint occurred. As illustrated in Figure 7, the deformation of the spine during a simulation is divided

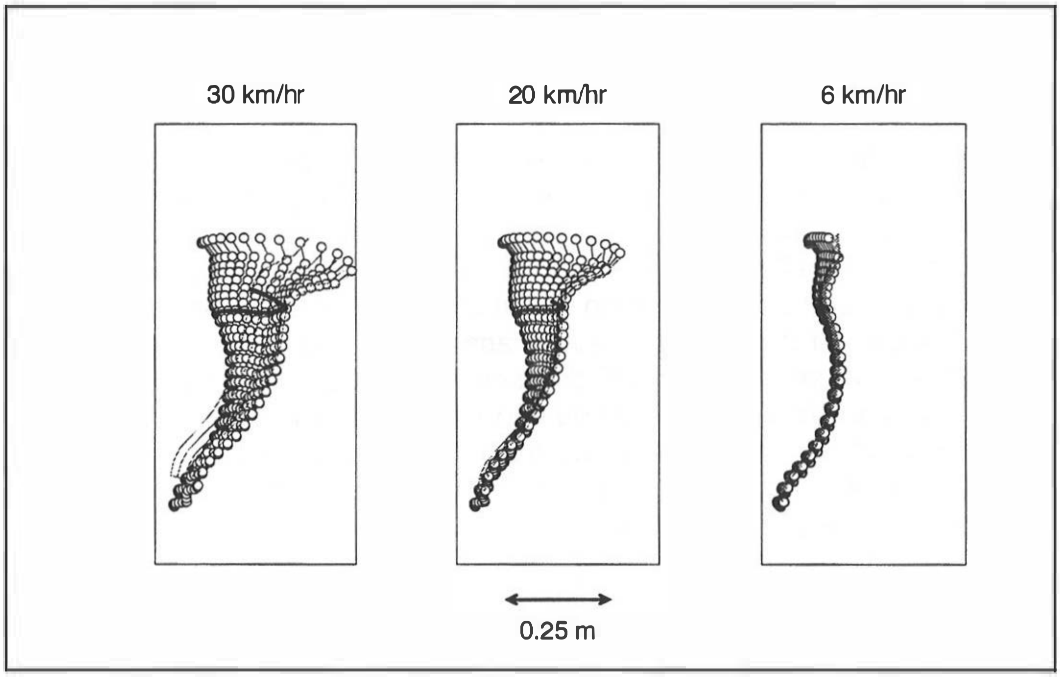


Fig. 7a - Changes in spine shape (without head restraint)

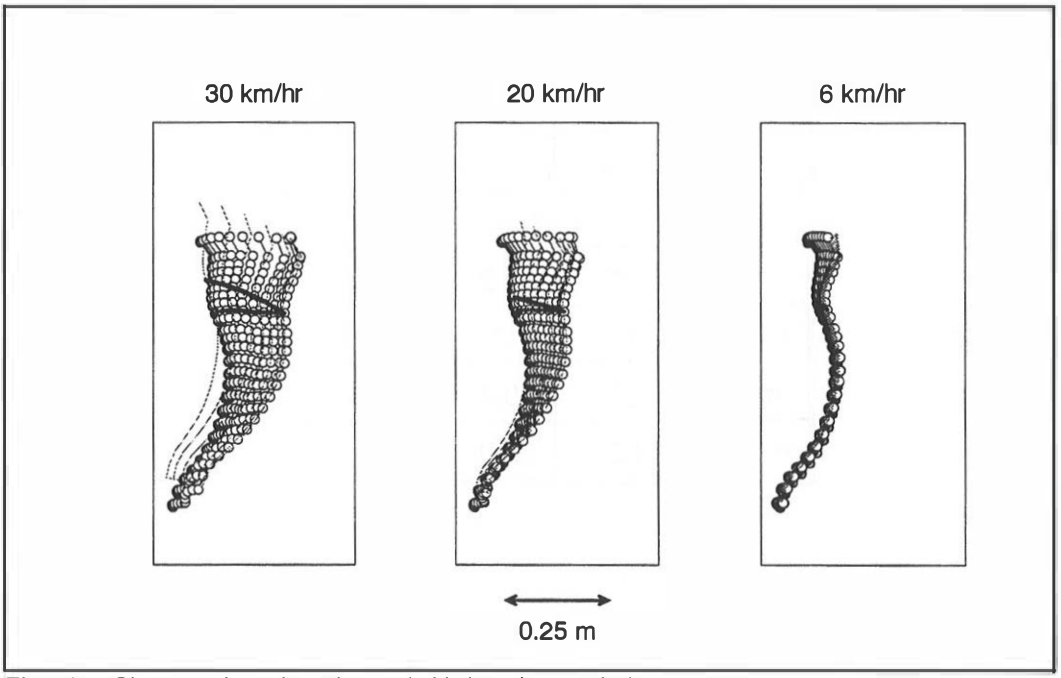


Fig. 7b - Changes in spine shape (with head restraint) into two phases: during the first phase (illustrated by the circles connected by solid lines in Figure 7), which will be called the "torso loading phase", T1 is moving backward with respect to the vehicle; the second phase (illustrated with thin dashed lines), the so-called "torso rebound phase", starts as soon T1 begins to move forward. A general observation is that during the torso loading phase, the spine is pushed back further as the level of severity of the pulses increases. Also, the head-neck motion is more pronounced in the cases without a head restraint compared to those with a head restraint.

Joint rotations - A comparison of the initial spine configuration with the final configuration during the "torso loading phase" reveals that extension occurs in the lumbar and thoracic spine, and that flexion takes place only at the mid-thoracic level. Also, the maximum extension occurs in the lumbar joints.

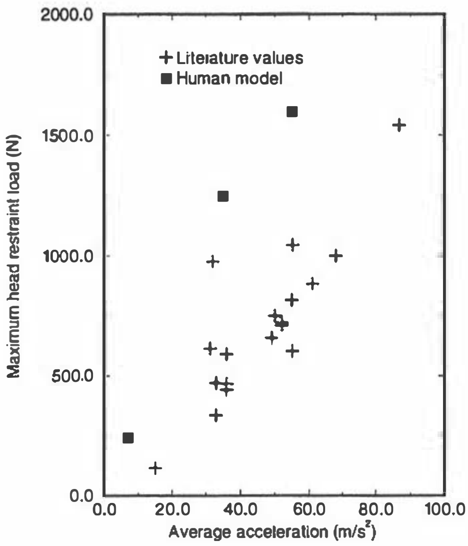
Axial displacements - The elongation of the spine (S1-T1) during the loading phase is negative (compression) and ranges between -3 mm for the low severity cases up to about -12 mm for the highest pulse (Table 4). For the medium and highest pulses, the elongation became positive during the torso rebound phase. No clear effect of the presence of a head-restraint on spine elongation could be detected.

Displacement of T1 - The vertical displacement of T1 during the torso loading phase (Δz_T1) ranges between 1 and 18 mm upward (Table 4). Δz_T1 is small for the low severity cases but larger for the higher pulses. Also, Δz_T1 was larger for the cases without a head restraint compared to those with a head restraint (Table 4). Finally, as can be seen in Figure 7, during the torso rebound phase the vertical position of T1 increased markedly, in particular in the higher severity cases.

Displacement of pelvis - The vertical displacement of the pelvis increases from about 2 mm in the low severity case to more than 5 cm for the higher pulse representing elevation from the seat (Table 4). An effect of the head restraint on the vertical pelvic displacement is not found.

Head rotation - The maximum head angle (defined as the angle between the Frankfort plane and the horizontal) is larger for the cases without a head restraint compared with those with a head restraint. Also, for the cases without a head restraint, the maximum head angle increased with the level of severity of the pulses (Table 4).

HEAD FORCES - The predictive ability of the model developed can be assessed by considering the head restraint contact forces and the shear forces at the OC joint. The predicted values for these parameters can be compared with values obtained from experiments with volunteers conducted by Mertz and Patrick (12). As expected, the maximum head restraint contact forces increase with the severity of the pulse, but are somewhat high compared to experimentally obtained values (Fig. 8). This is due to the fact that during the experiments the head was initially in contact with the head restraint, whereas in the simulations, there was an initial gap of about 7.5 cm. When the maximum head restraint contact force is plotted against the horizontal distance between the head and the head restraint, good correspondence with experimental data is obtained (Fig. 10). Regarding the OC shear forces, the maximum shear forces increased with the level of severity (Fig. 9). Also, the shear



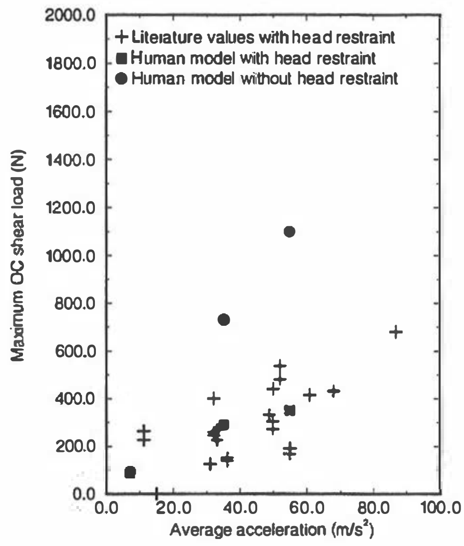
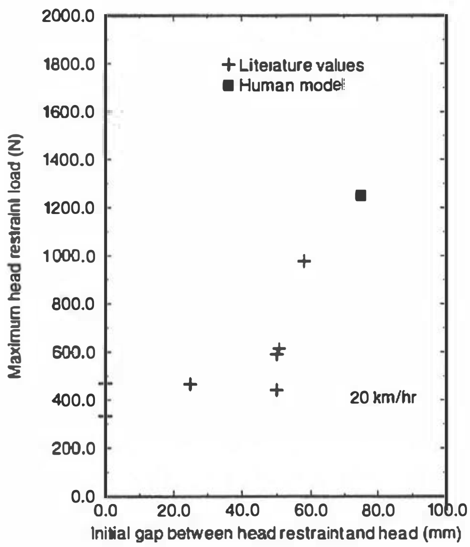


Fig. 8 - Head contact forces vs. sled acceleration

Fig. 9 - OC shear forces vs. sled acceleration



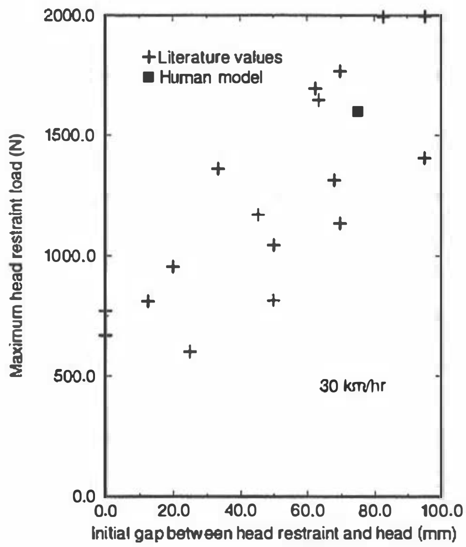


Fig. 10a - Head contact forces vs. horizontal distance to head restraint; 20 km/hr

Fig. 10b - Head contact forces vs. horizontal distance to head restraint; 30 km/hr loads were lower when a head restraint was used, compared with the cases when no head restraint was present, suggesting that head restraints reduce this type of loads. Finally, as illustrated in Figure 9, the predicted values for the OC shear forces correspond well with the experimental data.

Table 4 - Deformation of spine

case number	elongation (mm)	Δz_T1 total (mm)	Δz_pelvis (mm)	head angle (deg)
1 nhr_l	-3	1	2	30
2 nhr_m	-4	18	20	84
3 nhr_h	-11	14	51	92
4 hr_l	-3	1	2	30
5 hr_m	-8	7	22	14
6 hr_h	-12	5	52	11

DISCUSSION

In this study, a mathematical model of the human body was developed. Special attention was paid to the kinematics of the entire spine and the interaction with the seat. Given the short execution time for a simulation compared to a finite element model, the model developed is suitable for parametric studies to aid in a seat design process. Although the current study focused on the response in the mid-sagittal plane, and therefore only considered 2 degrees of freedom, in principle the complete model is three-dimensional and could be used for oblique impacts as well.

VALIDATION - A limitation of this study is that due to lack of experimental data from rear-end sled tests, the human model developed model could not be thoroughly validated. However, considering the overall kinematics of the model and comparing the head angle and the T1 displacements with experimental data, produced encouraging results. Despite the limitation mentioned above, this study represents a major improvement in the field of human body response modelling during rear-end collisions. However, to enable more complete validation and make this model more effective, more experimental data on rear-end sled test volunteers and cadavers is needed.

SPINE DEFORMATION - Using the generated data on the spine deformation, quantitative information on the relative contribution of different deformation mechanisms can de obtained. Vertical displacement of the T1 vertebral body can be caused by either rigid body motions of the spine as a whole (including the pelvis) and by deformation of the spine itself. The rigid motion of the spine is a result of pelvic translation and of pelvic rotation. Given the initial shape of the spine, forward rotation of the pelvis results in an upward movement of T1. The deformation of the spine itself consists of spinal elongation and joint rotations about their y-axes. From the results presented in this paper it can be concluded that the vertical displacement of T1 with respect to the vehicle is smaller than the vertical pelvic displacement (with respect to the vehicle) during the "torso loading phase", suggesting a downward motion of T1 with respect to the pelvis. The main mechanism for this "relative downward motion" of T1 appears to be the backward rotation of the spine in the joints at the lumbar level (lumbar extension). The compression of the spine itself is relatively slight, but accounts for up to 25% of the downward motion. The rotation of the pelvis does not contribute to this relative downward motion of T1 since it is usually forward, inducing an upward motion of T1.

CONCLUSIONS

- A mathematical model of the human body to be used for rear-end impact simulations is developed.

- Model validation is not complete due to lack of experimental data, but a comparison with existing data on overall body kinematics, maximum head angle, T1 displacements and head contact forces are encouraging. More experimental data on rear-end sled tests is needed to thoroughly validate the model developed.

- The vertical motion of T1 relative to the pelvis is downward. The main mechanism for this T1 motion is backward rotation in the lumbar spine joints, and to a lesser extend, spinal compression.

- Partly due to short execution time, the model developed is suitable for parametric studies to aid in a seat design process. Also, the model can easily be adapted into a 3-D version.

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