HUMAN BODY-CAR SEAT COUPLING UNDER REAR IMPACT

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ABSTRACT

The development of new protective systems must be performed on reliable tools and representative of alive human. In an earlier study a simplified and realistic model of the head-neck system under moderate rear impact was performed.

It is clear and often addressed in the literature that under such an impact configuration, the deformation of the torso and the car seat, is of extreme importance and defines the initial conditions of the head-neck system.

In order to address this issue, an original lumped model of the human torso was developed in the present study and coupled to a car seat-head rest complex. The hypothesis of linear behavior was used for the torso being subjected to small deformations . The modal analysis of the human torso in a seating position conduced by Kitazaki and Griffin in 1992 was used in this study for both masses and mechanical properties identification.

In order to reproduce the four mode shapes identified experimentally the torso was divided in six segments to obtain the five degrees of freedom with the head neck system. This model of minimum complexity but able to reproduce the 5 first experimental vibration modes was validated in the frequency domain in terms of natural frequencies and damping as well as mode shapes. In addition to the lumped approach, an external geometry was implemented in order to couple the human body model to a finite element model of the car seat also developed in the present studies. Rear impact simulations for the two different configurations (flexible and rigid torso) showed an increase of about 35% for the maximum T1 acceleration and an increase of about 65% for the acceleration slope when a rigid torso is considered. Realistic body behavior and accurate T1 acceleration are essential aspects in real world accident reconstruction as well as for seat-head rest evaluation and optimization against neck loading.

INTRODUCTION

Despite advances in safety devices, neck injuries in traffic accidents, especially non-severe rear impact accidents, continue to be a serious and costly social problem. The high cost of whiplash injury has been extensively documented in several countries (Szabo *et al* 2002 and 1996). The development of safety measures designed to decrease the incidence of whiplash injuries must be guided by meaningful and reliable human body surrogates. Most injury prevention strategies are based on impact analysis using anthropomorphic crash test dummies or mathematical models. Without proper evaluation of these experimental or computational models against the mechanical responses of the human body, it would not be possible to improve the current state-of-the-art neck injury prevention techniques. Unfortunately the spine is one of the most complex structures in the human skeletal system and its behavior during impact is still poorly understood.

At present there are no less than three crash test dummies dedicated for use in experimental rear impact analysis ; the Hybrid III dummy developed by Foster *et al* (1977), the BioRID II designed at Chalmers University Davidsson (1999) and the RID dummy proposed by TNO in the Netherlands Cappon *et al* (2001). A number of validation studies have been conducted on these dummies against volunteers and against post mortem subject neck responses (Davidson 1999, Cappon et al 2001, Davidson *et al* 1999, Prasad *et al* 1997, Seeman *et al* 1986 and Siegmund *et al* 2001) and have demonstrated the limited biofidelity of this human body surrogate under low speed rear impact. Optimization studies of the car-seat-head rest system were also described by Szabo *et al* 2002, Ishikawa *et al* 2000, Eichberger *et al* 1996 and Svensson *et al* 1993 and concluded that the safest protective system against whiplash depends on the dummy used.

The modeling of the human trunk began in the mid last century. Several kinds of models were developed either as a continuum or with lumped parameters. Most of these models do not have a realistic behavior compared to the human body. Either they are too detailed and involve a great quantity of not easily identifiable parameters with the existing experimental data, or they represent only one particular dynamic behavior of the trunk and cannot thus be used for other applications like simulations of rear impacts. Indeed, most of the spine studies developed characterize the global dynamic behavior of the trunk-head unit under seat ejection for military application. Typically, the models can be divided into two categories:

- The continuous models (Hess and Lombard 1958)
- The lumped models (Vulcan and King 1970,).

However, none of them have been studying the kinematics behavior of T1 under rear impact.

Many multi body system human model have been developed in the MADYMO software (TNO 1997) for rear-end impacts. Jernström *et al* (1993) presented a two-dimensional human model. Jakobsson *et al* (1994) compared the head angle of this model with that of a volunteer at Δv 8 km/h. The upper thoracic spine curvature of the model and the time span for the head to headrest contact were not in accordance with the volunteer response. Next, as tests are not easy to be performed on volunteers that only can be exposed to non-injurious impacts, Eriksson (2002) developed a very simplified three-dimensional model with mechanical properties tuned in order to fit the BioRID I dummy response Davidsson *et al*. 1999 integrating a flexible spine. Cappon *et al*. 2001 also developed a dummy in a whiplash project with a flexible thorax called RID 2.

Typically, numerical or physical spine model validation is conducted against volunteers or postmortem human subjects (PMHS) by comparing the evolution of recorded mechanical parameters over time with the human response. This methodology is limited as it is very difficult to characterize a multiple degrees of freedom system under impact in the temporal domain. These difficulties are well illustrated by the large number of test dummy evaluation and comparative studies found in the literature. The number of prototype versions and contradictions between study conclusions illustrate how difficult it is to explain some phenomena that are masked within the time domain. An other illustration can be found in Philippens *et al* 2002 study where “realistic” dummy head kinematics can be observed, but T1 accelerations were out of corridors. The reason for this is that the dummy response has to remain within ranges or corridors with wide tolerance. The evaluation process in the temporal domain is not sufficiently accurate to extract initial ramps, local peaks and oscillations that can be of great importance.

Despite this critical issue, recent researches in spine biomechanics have improved our knowledge of this complex structure. The limitations listed above illustrate the need for further experimental and theoretical analysis. The purpose of this paper is to apply modal analysis techniques to characterize the human trunk system in vivo and to develop a lumped parameters model of this segment in the sagittal plane.

Indeed modal analysis in engineering is non-destructive and used for identification of dynamic structures. In biomechanics the method has been used extensively for bone healing processing and for dynamic characterization of the human head (Hodgson *et al* 1967, Stalnaker *et al* 1971 and Willinger *et al* 1990). Contrarily to other studies, with respect to the spinal column, and in addition to impedance recording of a single degree of freedom, Kitazaki *et al* 1998 undertook a detailed experimental modal analysis of the whole column including the head. A total of 15 degrees of freedom were taken into account, 3 for the head, 10 for the spinal column and 2 for the frontal area of the abdomen. The seated subject was vibrated vertically. The transfer function in terms of the apparent mass between the input force and the different degrees of freedom accelerations enabled to extract the modal characteristics of the system in the modal domain, i.e. natural frequencies and eigen vectors or mode shapes. In this way, eleven vibration modes were identified between 1.8 Hz and 17 Hz due to back. The aim of Kitazaki's study was modal characterization and comfort. The modeling was therefore restricted to definition of the analytical transfer function rather than mechanical characterization of the human body.

In previous studies undertaken at ULP, the experimental modal analysis of the human head-neck system in vivo provided us with natural frequencies and mode shapes which constitute original validation parameters for dummy necks (Willinger and Bourdet 2002). A detailed description of the applied methodology can be found in Willinger and Bourdet 2004, demonstrating how this experiment provided the biomechanical background for dummy and numerical neck model evaluation (Meyer and Bourdet 2004).

In the present study, we used the results from Kitazaki *et al*. 1998 to identify a 5 degrees lumped torso model. In the following first section we will describe the modeling of the head-neck-torso unit and the identification methodology for the stiffness and damping parameters. We will then present the coupling of this model with a car seat model realistic boundary conditions to simulate low speed rear end impact.

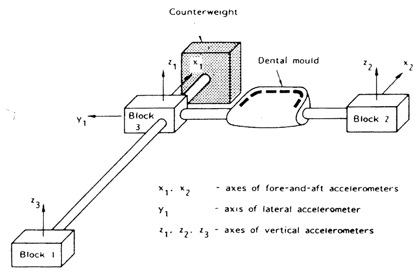
Finally a “standard” rear impact is simulated with the new torso model. The results are then compared to the response computed with a rigid torso under similar loading condition.

MODELING OF THE HUMAN TRUNK

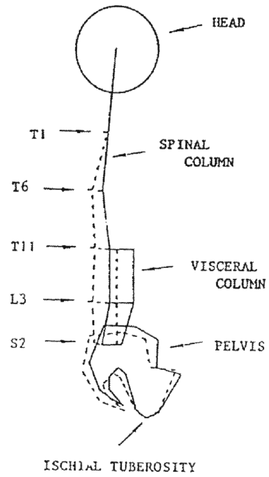
Experimental tests

The experimental data used in this study were completed in a context of ergonomics and comfort by Kitazaki *et al* 1998. The task was to characterize the movements of the head and the torso when the body was subjected to a vibration. The final aim of this research was to determine the frequency behaviors of the human body in order to better understand the origin of pains at the lumbar level. They also aimed to analyze the glance stabilization of a subject driving a car. After a first experimental attempt of modal analysis on the human body in vivo, Kitazaki *et al* (1998) decided to propose a modal analysis of the head-neck-trunk unit.

The studied system is shown in figure figure 1b. It is about the head-neck-trunk unit whose position in the sagittal plan is characterized by 15 degrees of freedom: the head T_x, T_y, T_z and θ_y recorded by an accelerometric device illustrated in figure 1a; and ten other sensors recording the accelerations of the five vertebrae T1, T6, T11, L3 and S2 in T_x and T_z.



(a)



(b)

Figure 1. (a) Measurement device for kinematics recording, (b) degrees of freedom of the human body for modal analysis [Kitazaki *et al* (1998)].

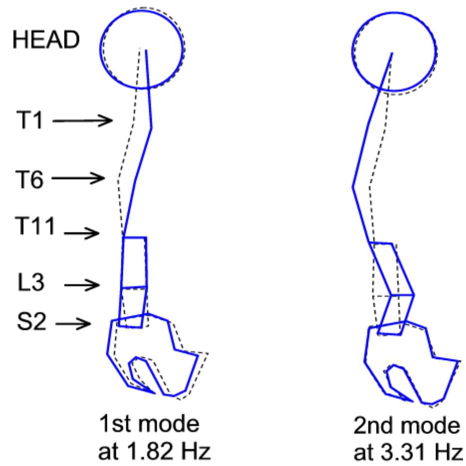
The system was excited by a vibratory platform recording the transmitted force and accelerations. The frequency exciter was able to transmit up to 10kN with a maximum displacement of 1m. The vibratory test consisted of a Gaussian random excitation *(Γ=1.7 ms^{-2} (rms), f=0,5 à 35 Hz*; during: 1 minute). Only one 32 years old male volunteer was subjected to the test. Thereafter, two types of experimental responses were analyzed: the transfer functions in terms of:

Apparent Mass : $A_{jk}=\dfrac{\Gamma_j}{F_k}$ and in terms of

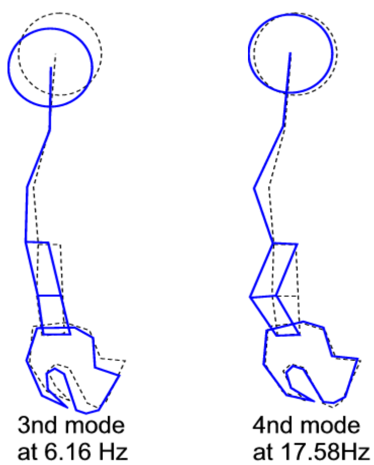
Transmissibilitie : $T_{jk}=\dfrac{\Gamma_j}{\Gamma_k}$

Where F_k et Γ_k are force and acceleration at the platform level (inputs).

This representation of the human body allowed the authors to write the transmissibility equations and to superimpose them with those recorded experimentally. The expression of the deformed mode shapes, illustrated in figure 2, and their quantitative description of table 1 was also described by this analytical transfer functions.



(a) (b)



(a) (b)

Figure 2. Representation of deformed mode shapes extracted by Kitazaki *et al* 1998, acordingly to the model.

While this study of high quality can be of very great interest in the analysis of car drivers' comfort, its applications in impact biomechanics are limited.

The limitations of this work, for a characterization of the spine column, are at two levels:

- the definition of the degrees of freedom is not adapted to a description of the cervical column,
- the analytical transfer function proposed not frequencies and modes shapes without mechanical parameters identification such as segment masses or rigidity and damping.

Four relevant deformed mode shapes are of interest for the spinal column modeling under rear end impact and will be considered in the following section.

Table 1. Quantitative results of the human body modal analysis.

Mode	*Natural frequency [Hz]*	*Damping ratio*
1	1.82	0.224
2	3.31	0.215
3	6.16	0.178
4	17.58	0.296

Lumped parameter model of the torso

In our study, we used the experimental data of Kitazaki (1992) and Kitazaki and Griffin (1998), to establish a minimum complexity lumped parameters model allowing the reproduction of a realistic dynamic behavior of the human torso. In order to obtain the deformed mode shapes given by Kitazaki, our model consists of five joints, as illustrated of figure 3. The head-neck joint remained blocked for this part of the study.

The model consisted of six segments respectively representing the lower and higher lumbar part, the lower and upper thorax, the neck and the head. Mass m_i and inertiae J_i from each part are concentrated at the gravity center G_i. Each joint has a stiffness k_i and damping c_i. We did approximate the angular functions to order 2 for all ψ_i angles around zero. The following functions are then obtained and reported in equation 1:

$$\sin(\psi_i)=\psi_i+O(\psi_i^2)$$
$$\cos(\psi_i)=1+O(\psi_i^2) \qquad (1)$$

The lengths, masses and inertiae were determined by anthropometric measurements and calculated using a geometrical model developed by Hanavan (1964). This model represents the human body by superimposition of ellipsoidal and cylindrical segments. The mass components are based on the regression equations reported by Clauser *et al* (1969).

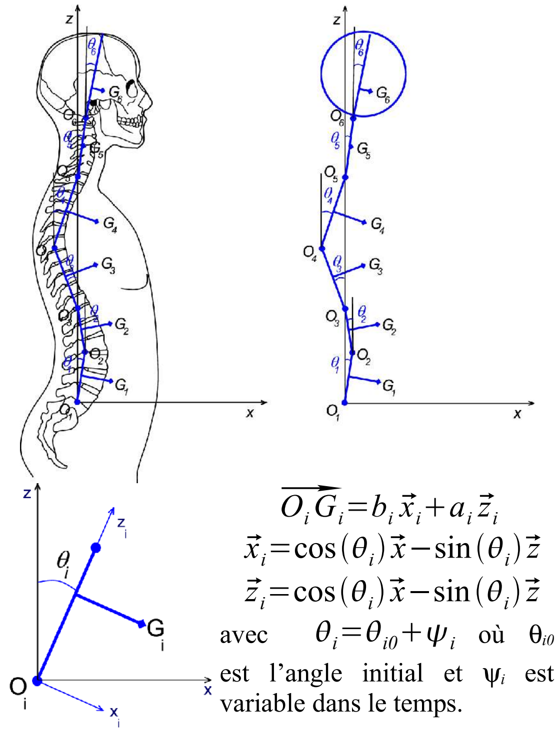


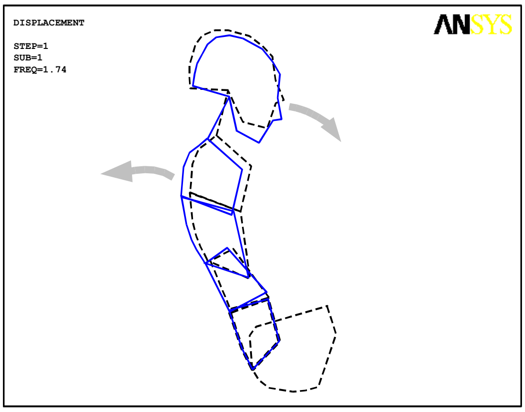
Figure 3. Representation of the lumped parameters model of the trunk.

In order to obtain the masses and inertiae of the five pivots model, we divided the trunk by an upper part and a lower part being the pelvic part. The values thus obtained, after having extracted from the literature the lengths necessary for calculation, are reported in table 2.

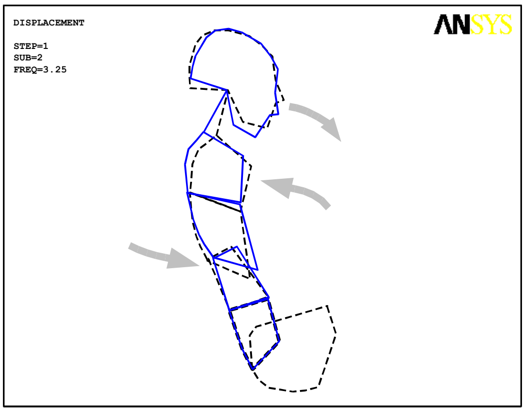
Table 2. Mass and inertial data of the trunk, the neck and the head.

Parts	*Mass [kg]*	*Inertiae /y [kg.m²]*
Lower Lumbar	3.6	0.01
Upper Lumbar	7.3	0.0281
Lower Torso	8	0.0352
Upper Torso	10.5	0.0603
Neck	1.7	0.002
Head	4.5	0.04

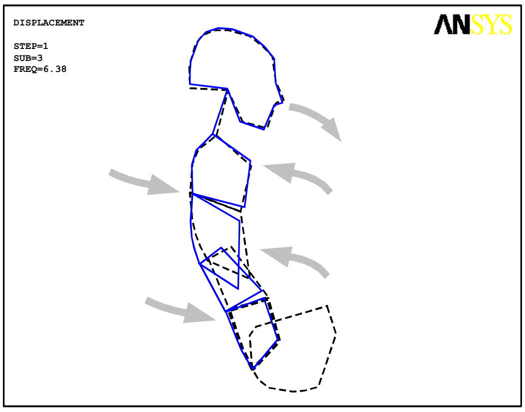
The lumped parameters model of the head-neck-trunk unit was introduced into the implicit finite element code ANSYS in order to calculate the natural frequencies and the deformed mode shapes of the system. The initial stiffness and damping values were selected so that the model presented the same deformed mode shapes as those obtained by Kitazaki at a similar natural frequency.



1st mode at 1.74 Hz

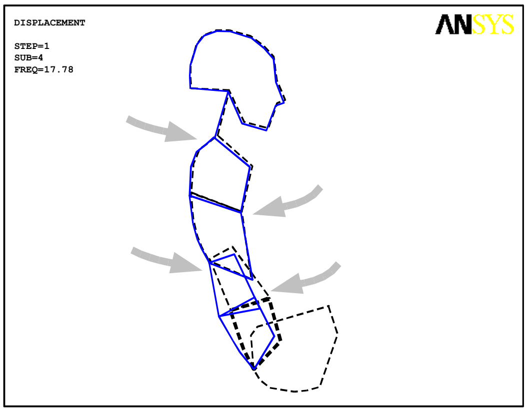


2nd mode at 3.25 Hz



3th mode at 6.38 Hz

Figure 4. Representation of the 3 first deformed mode shapes obtained under free vibration modal analysis with the proposed from the model.



4th mode at 17.78 Hz

Figure 5. Representation of the fourth deformed mode shapes obtained under free vibration modal analysis with the proposed from the model.

In order to be in the same configuration as in the experimental study provided by Kitazaki, we imposed a vertical displacement on all lower parts of the model including the legs and the feet. We also blocked the head-neck joint. Two types of analysis were carried out:

- •a free vibration modal analysis, which permitted to distinguish the various deformed mode shapes accordingly to the natural frequencies for an elastic behavior;
- a harmonic analysis which permitted to determine the true values of the natural frequencies and the damping ratios.

The free vibration modal analysis enabled us to determine four deformed mode shapes with natural frequencies over 1 Hz, illustrated in figure 4 and 5. This deformed modes shapes can be compared with those obtained by Kitazaki and Griffin (1998) reported in figure 2. In fact, another natural frequency appears at 0.38 Hz which corresponds to a deformed mode shape not reported by Kitazaki.

The four modes presented are considered as sufficient to validate the model as only stiffness and damping in S2, L3, T11 and T6 are to be identified. Indeed, we already identified the T1 joint stiffness during a previous study on modal analysis of the head-neck system (Willinger and Bourdet 2004).

A parameter optimization of stiffness and damping was then carried out on the model in order to obtain a good accordance of the natural frequencies and the damping ratio with those extracted by Kitazaki *et al* 1998.

A total of 27 iterations were necessary to obtain these parameter optimization. Results are reported in table 3 together with the experimental ones.

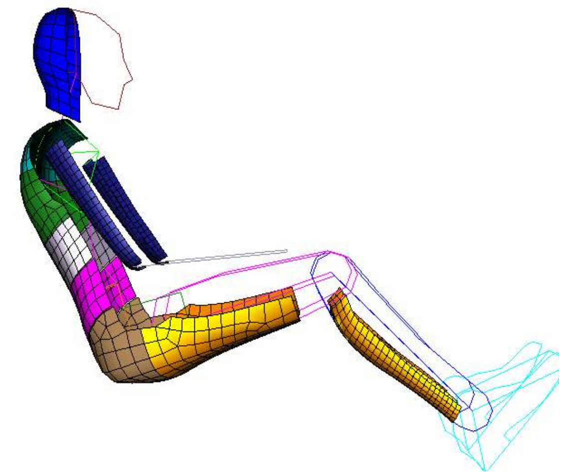
Table 3. Optimization model behavior compared with the experimental ones reported by Kitazaki *et al*.

Mode	***Natural frequency [Hz]***		***Damping ratio***	
	Exp.	Model	Exp.	Model
Mode 1	1.82	1.90	0.224	0.23
Mode 2	3.31	3.25	0.215	0.21
Mode 3	6.16	6.2	0.178	0.18
Mode 4	17.58	17.2	0.296	0.25

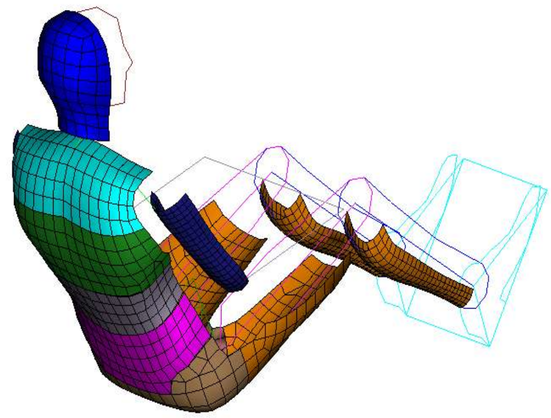
The model was then introduced into the explicit FE code RADIOSS (MECALOG). It is based on the lumped model presented previously. However the arms and the legs were added in order to take into account their the mass and inertial effects.

In order to reduce the number of elements, the model structure is defined with beam elements. An external shell representing the back of the car occupant was meshed and fitted to the human lumped model. The geometry of the surface is based on the volunteer's geometry by palpation of the back. Each segment is defined as a rigid body. Their mass and inertia are attached to the master node corresponding to the center of gravity of the considered torso part

Only surfaces in contact with the car seat were considered, as illustrated of figure 6. The considered surfaces are the following parts : torso surface (the upper thorax, the lower thorax, the upper and lower lumbar), the gluteal surface, the thighs, the legs, the arms and the head. The surface of the neck is related to the surface of the head. These considered surfaces were essential to carry out the coupling between human body and car seat, which is the subject of the following section.



(a)



(b)

Figure 6. Finite elements model of the human head-neck-trunk unit including surfaces in sagittal sight (a) and 3D sight (b).

CAR SEAT MODEL

The numerical modeling of a car seat aimed at giving realistic boundary conditions to the human model in the case of rear end impact. The car seat consists in various mechanical elements. The main parts of the seat were : the head-rest clamp, the head-rest foam , the foam of the backrest, the foam of seat base , the backrest spring, and the cover of the seat. The geometry of the seat was based on exiting car seat and is illustrated in figure 7.

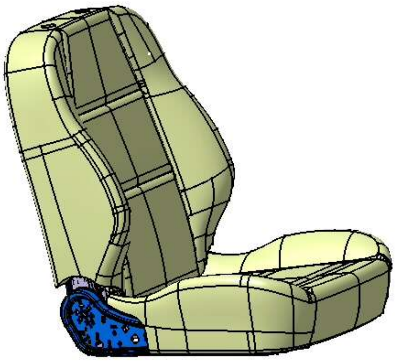


Figure 7. Representation of the car seat.

In this study, the material behavior laws for the foam and the cover were considered linear with material properties resulting from the literature.

The backrest was more detailed than base of the seat. Thus, a simplifying hypothesis consisted to model seat base with a flexible shell which aimed at limiting the movement of the thighs and pelvis. The mechanical properties of seat base have been extracted from a compression test and was considered with linear elastic shell elements and The Young modulus was of 1000MPa and the Poisson's ratio was 0.3. The thickness of the shell elements was of 1.5mm with a density of 500 g/l.

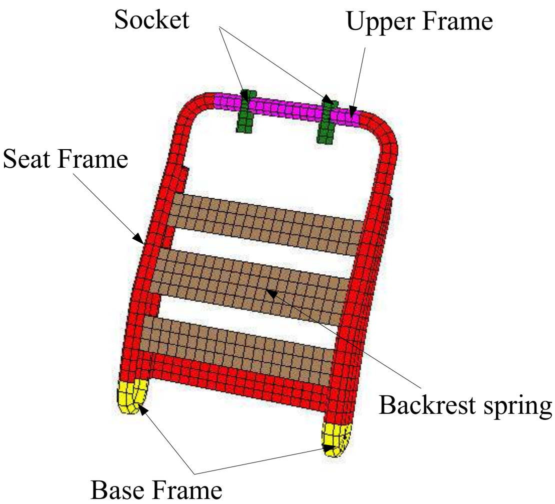


Figure 8. modeling of the backrest frame.

Special attention was paid to the backrest and headrest of the seat. The backrest frame was modeled with shell elements. The geometry was simplified as illustrated in figure 8. It was divided into three parts: the base frame considered as rigid body; the seat frame which can be deformable; and the upper frame also considered as a rigid body part. The seat base and the backrest frame were related by a spring fixed on the base frame. The sockets were fixed on the upper frame.

The real backrest spring consisted of metal wire connected to the seat frame. At the model level, simplification led in a three meshed bands with shell elements as illustrated of figure 8. The sockets were modeled with shell elements and were considered as rigid segments. They were connected with the upper frame by springs. In the same manner as for the seat base, the material properties of the backrest spring were determined in order to have a qualitatively realistic behavior under static loading. We thus obtained a Young's modulus of 230MPa and a Poisson's ratio of 0.29. The density was of 7.8 kg/l and the thickness of the bands were of 2mm. The stiffness of the springs between the sockets and the upper frame were chosen very high to be considered as rigid.

The backrest foam is modeled with 3D brick elements. In order to homogenize meshing, it was necessary to simplify its geometry. The mesh is presented in figure 9. The foam is divided into three parts (upper, medium and lower) which can have different mechanical properties. The mechanical properties were extracted from modal analysis on several samples of 100x100x40 mm3. The material behavior law used was linear elastic with a Young's modulus of 80kPa and a Poisson's ratio of 0. The density of the foam is given by the manufacturer to be 40 g/l.

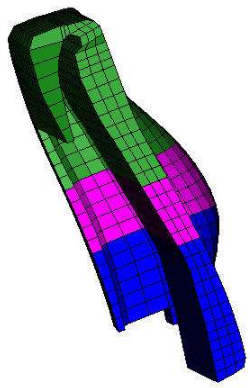


Figure 9. Modeling of the backrest foam.

In order to ensure numerical stability, a reinforcement shell was added on the back surface of the backrest foam. This is also covered with a fabric modeled with shell elements whose nodes coincide with those of the front external surface. The mechanical properties of the fabric have been extracted from static tensile tests. The value of the Young's modulus is then fixed at 1000MPa.

The head rest model consisted in the clamp and the headrest foam. The clamp is divided into two parts (figure 10): a deformable part which penetrates in the foam; and a rigid part which is related to the sockets with a very stiff spring. The foam is meshed with 3D brick elements. The mechanical properties have been extracted from modal analysis, in the same manner as for the backrest foam. Thus the Young modulus was 50kPa, the Poisson's ratio is 0 and the density was of 36 g/l. The head-rest foam was also covered with a fabric meshed with shell elements. The mechanical properties of the fabric are the same as those used for the fabric of the backrest.

Finally, the seat base and the backrest frame are bound by a spring at the level of the base frame whose stiffness is of 3300 kNm/rad. A particular attention was made on the junction of the upper frame level with the clamp of the head rest. The sockets are connected both with the clamp and the upper frame by a very stiff spring.

HUMAN BODY-SEAT COUPLING

The coupling of the human model with the seat model was done by adjusting both model in geometrical position of the H-point. In order to have an ideal contact at the beginning of impact, we have moved several nodes of the human model by giving to the column a curve adapted to that of the seat (figure 11a).

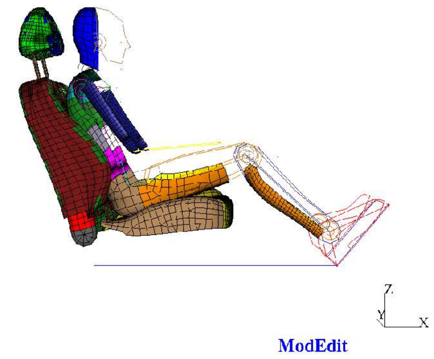
In order to simulate a rear end impact, we applied an acceleration pulse from a EuroNCAP type at 16 km/h to the seat, as illustrated in figure 11b. The results are shown in figure 12. The purpose of this simulations is to compare the T1 kinematics obtained with a rigid thorax and the flexible thorax developed in the present study.

Figure 10. Modeling of the headrest.

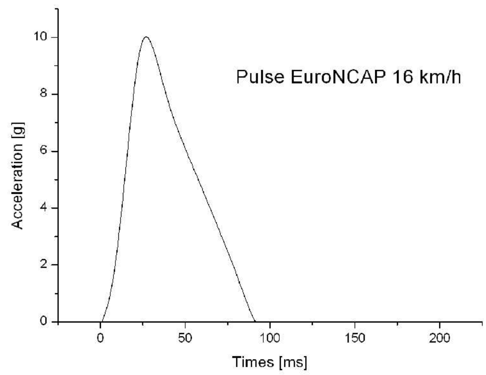
To compare the flexibility influence of the thorax, we have to simulate two types of rear impacts.

- a configuration with the torso model validated in the frequency domain called flexible torso;
- a configuration with the rigid thorso.

As represented in figures 13a and 13b, extracted at the same computing time (120 ms), an important differences in dynamic behavior can be observed with an amplified head extension in the case of a rigid thorax (figure 13b) whereas figure 13a shows a marked retraction movement.



(a)



(b)

Figure 11. Positioning of the human model in the seat model (a), Representation of the EuroNCAP pulse at 16 km/h applied to the human-seat unit model (b).

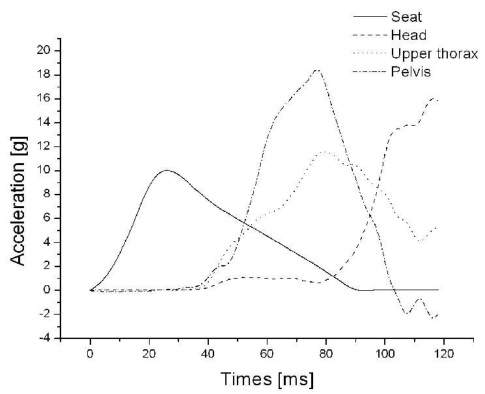
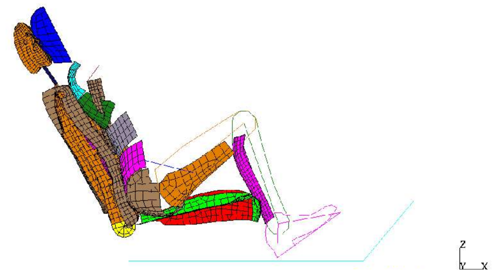
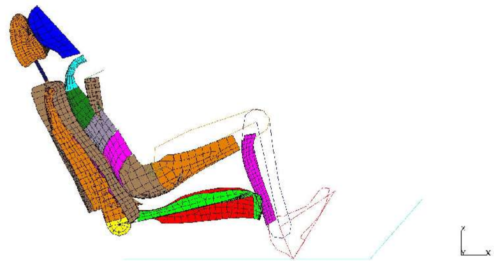


Figure 12. Results in terms of acceleration for several parts of the human-seat coupling model.



(a)

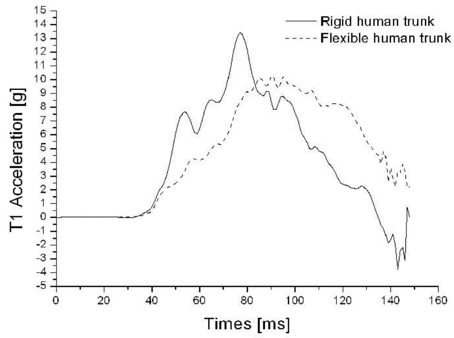


(b)

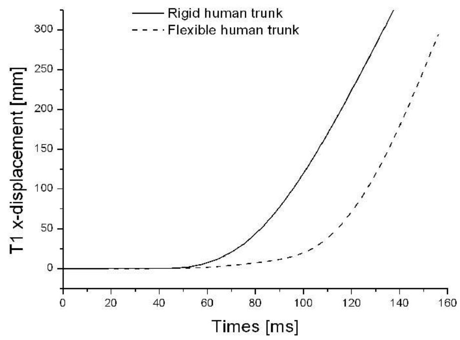
Figure 13. Simulation results under rear impact for a flexible column (a), and a rigid column(b).

Figure 14a shows the superimposition of the x-accelerations at T1 level, in the case of a rigid and flexible thorso. We clearly observe a difference of slope (62.5%) and amplitude (34.5%). Such a difference in T1 loading implies a radically different dynamic behavior of the head and neck, as shown in figures 15a, 15b and 15c for respectively C1-T1 relative displacement, head-torso angle rotation and C1-T1 relative velocity. The x-T1 displacement is more significant when the thorax is rigid, as illustrated in figure 14b. This is due to the fact that the backrest is loaded by all the trunk mass, while the borne mass by the backrest decreases with a flexible thorax. The same phenomenon is described in figure 14c for the Z-T1 displacement.

The effects of a various kinematics on the T1 level cause a relative displacement C1-T1 more significant for a rigid thorax (93 mm) than for a flexible thorax (77mm), as illustrated in figure 15a, with a gap of 20%. The relative velocity curves C1-T1 also shows a great variation up to a maximum of 45% (figure 15c). As for the head-torso relative rotation we can clearly see for the flexible thorax that the head has a retraction movement defined by the positive angles (figure 15b). On the other hand, a rigid trunk does not give the same behavior. Indeed, the head-torso relative rotation is first negative (extension movement) and then positive (retraction movement) caused by the head rest.



(a)



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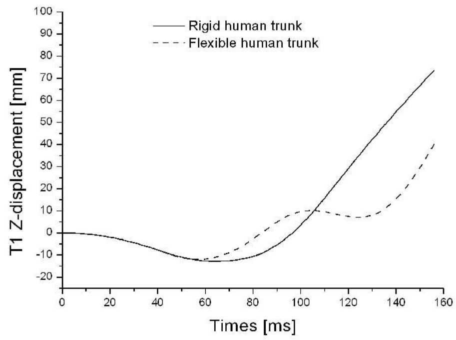

(c)

Figure 14. Superimposition of computed T1 acceleration for both rigid and flexible torso model under rear impact.

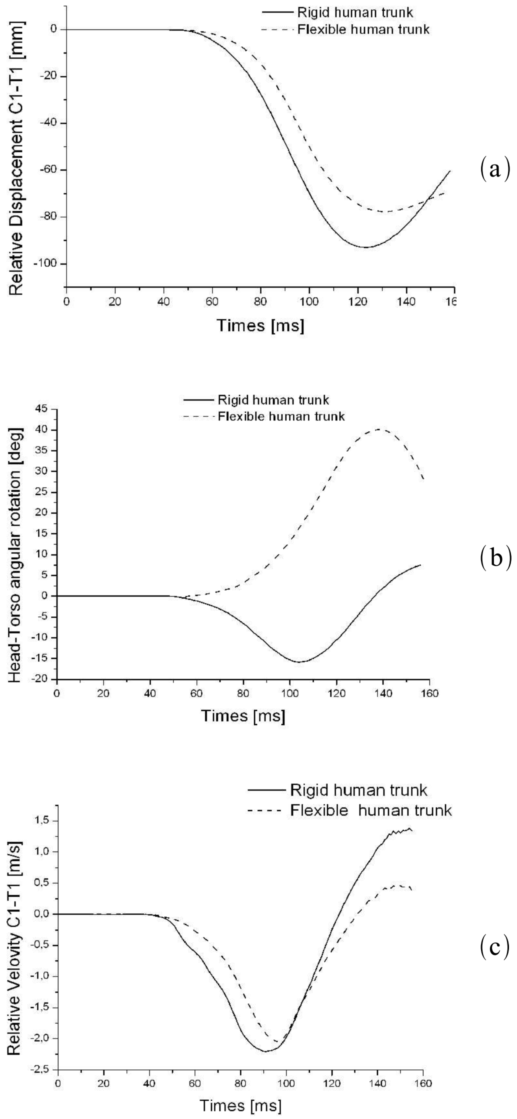

Figure 15. Superimposition of computed C1-T1 relative displacement (a), head-torso relative rotation (b) and C1-T1 relative velocity (c) computed with a rigid and non rigid torso model.

DISCUSSION

The discussion of this new human body-seat model is divided into two parts. The first one is the validation of the human torso model itself, and the second deals with the results at T1 level, the main parameters of the head-neck loading.

Most of the studies of the spine characterization were conducted in terms of intervertebral loading and kinematics of some vertebrae in the temporal domain (Kroell *et al* 1974, Stalnaker *et al* 1971 and Viano *et al* 1989). Kitazaki *et al* 1998 applied modal analysis technic and extracted deformed modes which four of them correspond to the spinal column deformations. The linearity of the system was checked with the coherence function which remained close to 1. The superimposition of the numerical model analysis with the experimental results made it possible to define the stiffness and damping parameters for each joints of the torso model. Even if a more realistic modeling of the human torso behavior is proposed in the present study, it must be mentioned that the model validation is limited to the sagittal plane and based on one single 38 year old human male volunteer. Further analysis including female is therefore needed.

A number of validation and comparative studies of rear impact dummies are reported in the literature (Cappon *et al* 2001, Kim *et al* 2001, Siegmund *et al* 2001). All of them were conducted in the time domain. The main improvement observed using rear impact dummies was a more flexible spine than for Hybrid III dummy. Two recent comparative studies (Prasad *et al* 1997 and Philippens *et al* 2002) demonstrated that BioRID and RID2 had very similar responses under moderate impact although BioRID has a flexible thorax. Prasad *et al* 1997 concluded that Hybrid III is suitable for rear impact testing in the 8-24 km/h range when Philippens *et al* 2002 had the opposite position. This can be explained by a not enough accurate model evaluation in the time domain. In fact, The models are validated against volunteer and cadavers kinematics in the time domain in terms of corridors. This kind of validation is not accurate enough to extract all the dynamic behavior of the torso. Other contradictions were obtained in the time frame when Philippens *et al* 2002 found that for rear impact dummies head kinematics were acceptable whereas T1 kinematics were not. It is questionable here how the head can behave accurately when T1 does not, given that T1 is the input of the head-neck loading. In addition to the difficulty related to analyze in the time domain, authors often add complexity by considering seat and thorax effect to the neck validation. This is illustrated by Kim *et al* 2001 and Szabo *et al* 2002.

At the theoretical level Eriksson *et al* 2004 recently proposed a torso-seat coupling through a MADYMO BioRID I model coupled to a simplified seat model. The purpose was to reconstruct real world rear impacts and no in deep validation of the human torso was addressed in this study.

In our study we showed that a flexible thorax gave clearly different T1 responses compared to a rigid thorax. These boundary conditions applied to the head-neck system changes drastically the results in terms of head acceleration, head-neck relative rotation or displacement.

CONCLUSION

The experimental data extracted from the modal analysis of the human torso by Kitazaki *et al* 1998 enabled us to define a lumped model of the human torso with five degrees of freedom. This model is then able to reproduce the natural frequencies and deformed mode shapes of the column extracted by the preceding authors. Coupled with a car seat model, it can thus be used to simulate low speed rear end impact more realistically.

A comparative study contributed to show the influence of the thorax flexibility. The boundary conditions of the head-neck unit, imposed by T1 kinematics, showed a very different dynamic behavior of the head and neck when a flexible or rigide torso was considered. It then becomes very significant to have a realistic modeling of the dynamic behavior of the column, if we want to improve the protection systems for a car occupant under low speed rear end impact

In a further development it will be possible to conduce a parametric study on seat characteristics and optimize the seat against the biomechanical response of the human torso-neck-head complex.

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