FINITE ELEMENT MODEL OF THE HUMAN THORAX VALIDATED IN FRONTAL, OBLIQUE AND LATERAL IMPACTS : A TOOL TO EVALUATE NEW RESTRAINT SYSTEMS

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A 3D model of the thorax including shoulders and abdomen was developed. The model was designed to predict human responses and kinematics under all crash conditions (frontal, oblique and lateral). The geometry chosen is that of the 50th percentile adult male in a seated posture. The anatomy is represented by 6000 spring, shell, membrane and brick elements. 22 validation test configurations were carefully selected and 50 validation corridors were established. There is good agreement between model results and experimental data so the model is already in use for several applications such as the development of cadaver experiment protocols.

IN RECENT YEARS, a considerable increase in the use of numerical simulation has been noted in dealing with the problems associated with car safety and with the protection of occupants. Recently, models of standardised crash dummies have been developed (Midoum 1993, Fountain 1996). They undoubtedly contribute to the development of improved restraint systems, but they suffer from the defects of the physical models from which they have been derived, particularly concerning their limited biofidelity. It is for this reason that several teams are working on the development of numerical substitutes for the human being (Huang 1994, Planck 1994).

The "Laboratoire d'Accidentologie et de Biomécanique" has been conducting research in this field for several years. While studies were carried out on the pelvis (Renaudin 1993) and on the neck (Dauvilliers 1994), particular attention was paid to the human trunk, which is often the segment that suffers serious injuries in road accidents.

The work consists of making a finite element model of the thorax, the abdomen and the shoulders, and validating it with respect to a large database of cadaver test results. The result is a biofidelic and omnidirectional model of the human, that is, it can simulate the human thorax response to impacts in all directions. As of now, this model is capable of predicting more than just kinematics, its responses in terms of physical magnitudes translating the global deformations (impact load, deflection, etc...) and of the main lesional criteria are well correlated with those of the subjects tested.

OBJECTIVE AND METHODOLOGY

The objective of this project was to develop a finite element model of the seated 50^{th} percentile adult male.

Principal specifications :

- Radioss software
- limitation of the processing time step (1μs)
- limitation of the maximum number of elements (6000)
- correct representation of the anatomy
- prediction of responses and kinematics in all impact directions and across the whole range of severities found in automobile crashes.

For this model, there was no requirement to fully validate the model's capability to predict the known injury criteria neither to develop new criteria.

In order to achieve the validation objectives, a large database from global tests on fresh cadavers was established. We have selected each configuration attentively while trying to obtain the widest range of different test conditions. These experimental data have been synthesised in the form of behavioural corridors. The geometry of the average occupant has been defined and a homogeneous mesh of the whole thorax has been made. A literature study provides information on the ranges of variation of the model's parameters. The calibration was made with respect to four priority corridors (two speeds and three impact directions), the responses of the model were then compared to the whole of the validation database.

VALIDATION TEST DATABASE

The database was constituted from a selection of global tests carried out on only fresh cadavers. Test configurations were selected according to experimental protocols, the accuracy of the boundary conditions and the nature of the accessible results. In several cases, the accelerometer data and the films were obtained and the analyses could be made completely in the laboratory. No method of normalisation was used and the raw data were taken into account in every case. The objective was to have the model responses evaluated with respect to the greatest number of tests, in this way the responses of the average occupant can be approached. The physical magnitudes which characterise the responses and the kinematics and which are reproducible from one test to another were used. We made a selection of the subjects tested in order to obtain the most pertinent and, therefore, the narrowest possible corridors. Subjects with atypical behaviours (excessive fragility, corpulence too far removed from the average, etc.) within the populations tested were eliminated. For each configuration and for each severity, corridors were established by choosing the envelope formed by the minimum and maximum of the selected curves.

Two types of test were selected : Impactor tests and belt compression tests.

IMPACTOR TESTS - All the test configurations selected are describe in Table 1, please refer to the reference list for a precise description of the experiments.

Table 1 : Description of the impactor tests

(The letter and the figure in parenthesis refer to notes following the table)

configuration	author	direction	position subjects	type impactor	target point	mass kg	speed m/s	measure of deflection	total n° subjects (j)	N° select. subjects (k)	corridors (l)
thorax1	Bouquet	frontal	sitting	guided	(f1)	23.4	3.4	(i1)	4	3(k1)	a-f
thorax2	Bouquet	frontal	sitting	guided	(f1)	23.4	5.8	(i1)	4	3(k2)	a-f
thorax3	Kroell	frontal	sitting	guided	(f1)	23	4.9	(i1)	-	4(k3)	a-c
thorax4	Kroell	frontal	sitting	guided	(f1)	23.4	6.9	(i1)	-	7(k4)	a-c
thorax5	Kroell	frontal	sitting	guided	(f1)	22.2	9.9	(i1)	-	5(k5)	a-c
thorax6	Kroell	frontal	(d1)	guided	(f1)	10.4	7	(i1)	6	6(k6)	c
thorax7	Stalnaker	frontal	sitting	guided	(f1)	10	6	(i1)	9	9(k7)	c
thorax8	Bouquet	lateral	sitting	guided	(f2)	23.4	3.3	(i2)	6	3(k8)	a-f
thorax9	Bouquet	lateral	sitting	guided	(f2)	23.4	5.9	(i2)	4	3(k9)	a-f
thorax10	Viano	oblique	standing	pendulum	(f3)	23.4	4.4	(i3)	5	5(k10)	b,c
thorax11	Viano	oblique	standing	pendulum	(f3)	23.4	6.5	(i3)	5	5(k11)	b,c
thorax12	Viano	oblique	standing	pendulum	(f3)	23.4	9.3	(i3)	5	5(k12)	b,c
shoulder1	Meyer	lateral	sitting	guided	(f4)	23.4	5.5	-	7	7(k13)	b
abdomen1	Viano	oblique	standing	pendulum	(f5)	23.4	4.8	(i4)	6	6(k14)	d,c
abdomen2	Viano	oblique	standing	pendulum	(f5)	23.4	6.8	(i4)	4	4(k15)	b,c
abdomen3	Viano	oblique	standing	pendulum	(f5)	23.4	9.4	(i4)	4	4(k16)	b,c

Table 1 notes :

- thorax1 : configuration 1 for the thorax, shoulder1 : configuration 1 for the shoulder, abdomen1 : configuration 1 for the abdomen
- thorax1, 2, 8 and 9: in-house analyse (Bouquet 1994), thorax3-6: (Kroell 1970, 1974), thorax7: (Stalnaker 1973, 1974), thorax10-12, abdomen1-3 (Viano 1989), shoulder1 (Meyer 1994)
- (d1): subjects sitting with rigid backrest
- (f1): 4^{th}. Intercostal space, (f2): centred on the thorax in the alignment of T8/T9, (f3): 30° rotation of the body, impact centred on the xiphoïde, (f4): head of humerus, (f5): 30° rotation of the body, centred on the abdomen, 7.5 cm under the xiphoïde
- (i1): thoracic frontal deflection, (i2): ½ thorax deflection, (i3): 30° whole thoracic deflection, (i4): 30° whole abdomen deflection
- (j) : total number of cadavers in the test configuration selected
- (k) : number of cadavers selected and names of the cadaver given by the authors of the tests. (k1): MRS1,MRS2,MRS3; (k2): MRS2, MRS4, MRS6; (k3): 42FM, 53FM, 60FM, 45FM; (k4): 12FF, 14FF, 15FM, 18FM, 19FM, 20FM, 22FM; (k5): 24FM, 31FM, 32FM, 37FM, 55FF; (k6): 48FM, 50FM, 51FM, 52FM, 56FM, 58FM; (k7): 11M, 15M, 16M, 17M, 21M, 22M, 23M, 18F, 20F; (k8): MRL2, MRL6, MRL8; (k9): MRL3, MRL7, MRL9; (k10): 17, 29, 36, 40, 41; (k11): 4, 5, 7, 9, 11; (k12): 3, 14, 18, 33, 37; (k13): EPM1-7; (k14): 19, 23, 24, 30, 42, 43; (k15): 6, 8, 10, 12; (k16): 20, 28, 34, 15.
- (l) : Corridors built for each configuration, the definition of the corridors is the following one : a: deflection versus time; b: impactor load versus time; c: impactor load versus deflection: d, e, f: respectively impactor, vertebrae T1 and vertebrae T12 displacement in the direction of the impact versus time

BELT COMPRESSION TESTS - For these tests (Cesari 1990, Bouquet 1994), the subjects are laying supine on a rigid, flat table, a seat belt strap passes diagonally across the torso. Via a system of rods, an impactor launched with a certain velocity, loads the two extremities of the belt.

Three impact energies are selected:

- configuration A : M=22.4kg, V=2.9m/s

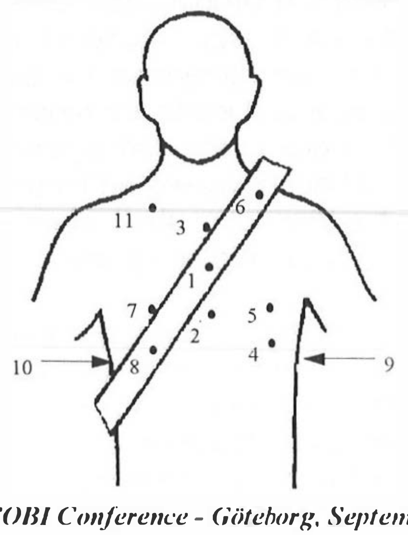
names of the selected tests : THC64, THC68, THC76, THC78, THC90, THC92

- configuration B : M=22.4kg, V=7.8m/s

names of the selected tests : THC11, THC12, THC14, THC15, THC16, THC17

- configuration C : M=76.1kg, V=2.9m/s

names of the selected tests : THC65, THC69, THC75, THC77, THC91, THC93



The deflection is measured vertically at ten points on the thorax and horizontally at one point (N°10), but only nine points have been taken into consideration (point 9 and 11 were not carried out for every tests): N°1: mid sternum, N°2: lower sternum, N°3: upper sternum, N°4: 8^{th} rib left, N°5: 5^{th} rib left, N°6: mid clavicle left, N°7: 5^{th} rib right, N°8: 8^{th} rib right, N°10: lateral 8^{th} rib right.

Figure 1 : View of the target positions

In total, the database is made up of a selection of 97 cadaver tests and 78 validation corridors.

FINITE ELEMENT MODEL

GEOMETRY - The geometry of the model comes mainly from two sources of anthropometric data. The first, by Robbins (1983), gives the external surface points and the locations of some bones of a median male in the driving position. The positions of the bones were obtained by palpation, consequently the precision of the measurement is poor. The second source is a set of centimetric scanner sections of a lying individual, the subject is a 50^{th} percentile male in size and in weight. Other anthropometric and anatomic studies were consulted (Dansereau 1988, Rebiffé 1982, Maertens) to verify the coherence between the two primary sources of data and to find, for example, the inclination of the ribs and the curvature of the spinal column.

MESH - In order to respect the constraints concerning the number of elements and the time step, an element size of the order of one to two centimeters is used. This limitation imposes a certain number of choices in the modelling.

Internal organs - All of the soft tissues filling the ribcage as well as the abdominal organs, are represented in a global manner by brick elements. The different thoracic-abdominal organs (lungs, heart, liver etc.) are not described geometrically because of the lack of information on their mechanical characteristics and on the manner in which they are linked to one another. At the beginning of the project, we also wanted to use a single material for all the internal organs. However, the experimental curves reveal different stiffnesses in the internal organ regions, for instance between the hard thorax and the abdomen (Viano 1989). Thus, in a premilinary approximation we decided to attribute different mechanical characteristics to three regions which roughly correspond to the lungs and the heart for the first region, to the spleen and stomach for the second one and to the intestines for the third one.

Spinal column - The thoracic and lumbar portions of the spine are represented by a set of brick elements. The geometry of each vertebra is reduced to that of its vertebral body, modelled by a rigid body. The intervertebral joints are reproduced, in a global manner, by brick elements which represent the intervertebral disks.

Ribcage - The ribcage is made up of ribs, intercostal ligaments, sternum and costal cartilage. These different parts are represented by shell elements which cover the "internal organ" bricks. In the context of limiting the propagation of hourglass modes, the shells used are triangular except for the intercostal ligaments, for which they are quadrangular. These different tissues are characterised by their shell thicknesses and mechanical properties. The costal-vertebral joints are modelled by a direct connection from the extremity of the rib to the edge of the corresponding rigid vertebra. To represent the influence of the abdominal aponevroses and muscles, the surface of the ribcage is extended downwards by triangular shell elements with characteristics close to those of the ligaments. These elements also have the advantage to limit hourglass problems.

Muscles and adipose tissues - The muscles and adipose tissues covering the ribcage are represented by a layer of brick elements connected directly to the shells of the ribcage. The volume of the shoulders is meshed with brick elements.

Skin - The skin of both the thorax and the shoulders is meshed in quadrangular membrane elements directly covering the "muscle" bricks.

The whole of the previously described part of the model forms a homogenous mesh without discontinuity.

Shoulder articulation complex - The shoulder complex is made up of three bones : the clavicle, the head of the humerus and the scapula. A square-sectioned cylinder, described by quadrangular shells, represents the clavicle. This cylinder is deformable, its two extremities are rigid. The head of the humerus and the scapula are modelled as rigid bodies. Because of the anatomical complexities, the nodes of the three bones of the shoulder are not coincident with those of the "muscle" bricks. The connection is made for the clavicle by using interfaces, and for the head of the humerus and the scapula by the inclusion of "muscle" nodes in the rigid bodies of these two bones. Three dimensional springs model the joints between the sternum and the clavicle, the clavicle and the scapula and the scapula and the humerus. The scapula is linked to the spine by a layer of shell elements interfaced with the surface of the rib cage.

MATERIALS, MECHANICAL PROPERTIES, DENSITIES - The literature was consulted to obtain the mechanical characteristics of the model in terms of recognised values and of the associated ranges of variation. Information concerning the solid materials such as the bones is accessible, but it is much more difficult to obtain such information, particularly dynamic, on the soft tissues. Furthermore, the modellings and the hypotheses used must be correctly taken into account in the choice of mechanical characteristics.

Bones of the ribs and of the clavicle - The assimilation of the ribs to shell elements of constant thickness is a significant approximation : an elliptical section with two types of bony tissue (cortical and spongy) is represented by a rectangular homogeneous section. The estimation of the mechanical characteristics and of the thickness of these elements is consequently a delicate operation. The shell thickness value chosen is 3mm and the fine tuning of the rib-cage model was performed with this value. However, reconsidering the average proportion of cortical bone in a rib section given by Got (1975) and the irregularities of the distribution of this quantity, a thickness value of 2mm might have been preferable. The preceding point and the fact that the rib model values were tuned in order to obtain good responses of the ribcage at different severity levels, even in the case of fracture, explain the low Young's modulus value chosen. During the development of this part of the model, which is characterised by a juxtaposition of materials with significant stiffness differences, we have to cope with numerical problems such as apparition of hourglass modes and lack of stability.

Soft tissues - The soft tissues are modelled by brick elements, their behaviour is associated with a viscoelastic law (VEL) defined by a modulus of compressibility K and a function of relaxation in shear : $G(t)=G0+(G0-GI)*\exp(-\beta t)$

where : G0 = short term shear modulus

GI = long term shear modulus

β = decay constant

The reasons for this choice are, first, a translation of a physical phenomenon and secondly, a way to limit the appearance of the hourglass modes. The coefficients selected for these viscoelastic laws nevertheless tend to reproduce an elastic behaviour. More complex material laws have not been used because of the limitation of the calculation time.

Internal organs - To represent the behaviour of these tissues, Plank (1994) uses a VEL (K=0.287MPa, G0=0.059Mpa, K=0.0024Mpa and β=100). Huang (1994) uses an elastic material of low stiffness (E=0.0084MPa, ν=0.47) backed up by a set of damping elements (c=10Ns/m). Sundaram (1977) proposes values of E ranging from 0.017MPa to 0.02MPa. We used a VEL with parameter values in the same order of those presented and we fine tuned the three groups of coefficients by comparing the model responses to the four selected corridors (cf RESULTS).

Muscles - Plank uses a VEL (K=0.23MPa, G0=0.07Mpa, K=0.024Mpa and β =100). On the other hand, Huang (1994) assimilates the muscles to an elastic material (E=20MPa, ν=0.3). According to the authors, the values of Young's modulus can vary over a very large range (E=0.07MPa-2650MPa).

Intervertebral disks - For the intervertebral disks, Huang (1994) has chosen the same values as Plank (1994) (E=10MPa, ν=0.2 ane ρ=2.7g/cm^3). These values have also been kept.

Costal cartilage - For these tissues, Huang (1994) proposes an elastic material (E=0.003MPa, ν=0.42 and ρ=2.7g/cm^3) whilst Plank (1994) uses higher stiffness values(E=10.3MPa, ν=0.3 and ρ=1g/cm^3).

Intercostals tissus - The values of Young's modulus vary from 20MPa to 3000MPa according to the different models.

Table 2 : Values selected for the model

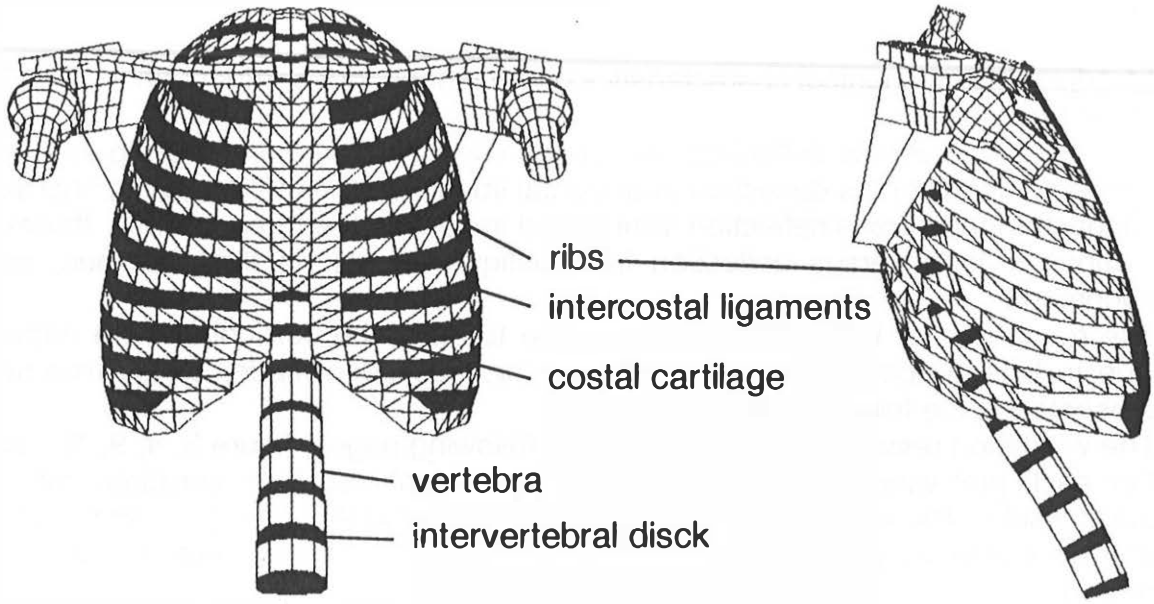
	element	thickness	E	K	G0	GI	β	ν	ρ
units	-	mm	MPa	MPa	MPa	MPa	1/ms	-	g/cm^3
bone	shell3-4	3	2500	-	-	-	-	0.3	1.8
Organs 1	brick	-	-	0.066	0.014	0.0107	1	-	1
Organs 2	brick	-	-	0.25	0.054	0.04	1	-	1
Organs 3	brick		-	0.166	0.036	0.027	1	-	1.1
disk	brick	-	-	16.7	3.57	2.68	1	-	1.1
cartilage	shell3-4	3	15	-	-	-	-	0.4	1.2
intercost	shell4	3	100	-	-	-	-	0.4	1.1
muscles	brick	-	-	0.25	0.115	0.086	1	-	1
muscles	shell4	1	300	-	-	-	-	0.3	1.1
Visc skin	shell3	1	5	-	-	-	-	0.4	1.2
skin	shell3-4	1	10	-	-	-	-	0.3	1.1

organs 1 : upper thoracic organs, organs 2 : lower thoracic organs, organs 3 : abdominal organs.

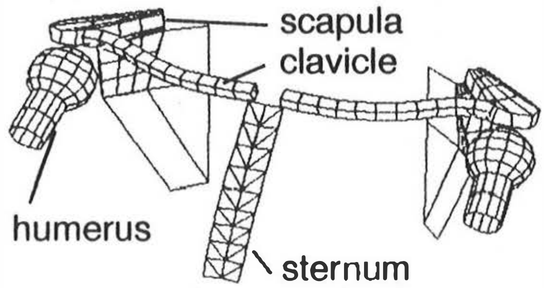
INTERFACES - An interface is defined in order to make sliding possible between the layer of membrane elements linking the scapula to the spinal column and, the rib cage. Five interfaces are defined to compensate for the non-coincidence of the meshes of shoulder bones and of the thorax in the assembly (clavicle/skin, head of the humerus/skin, scapula/skin, clavicle/ribs, scapula/ribs). Finally, two interfaces are put into place to limit certain deformations during very violent impacts. These problems arise from the use of behaviour laws which are not costly in terms of calculation time for the soft tissues, but which are insufficiently rigid for high compressions. These interfaces come into play when the two surfaces are abnormally close to one another (ribs/skin, sternum/vertebrae).

INERTIA / MASSES - The density parameters come from the literature, the overall masses and inertias have been adjusted with respect to the data of McConville.

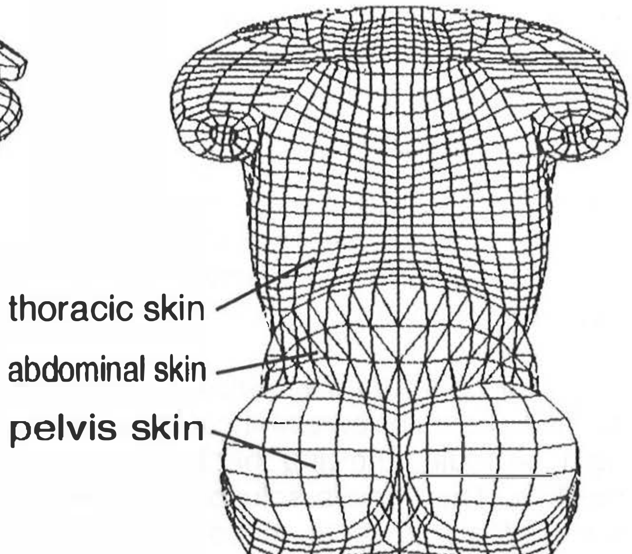
BOUNDARY CONDITIONS - The head and the neck are not represented, their influence on the thorax can be neglected at least at the beginning of most of impacts. The upper and lower limbs are represented through articulated rigid bodies, the pelvis region is rigid. The mass and the inertia of these segments have been precisely taken into account.



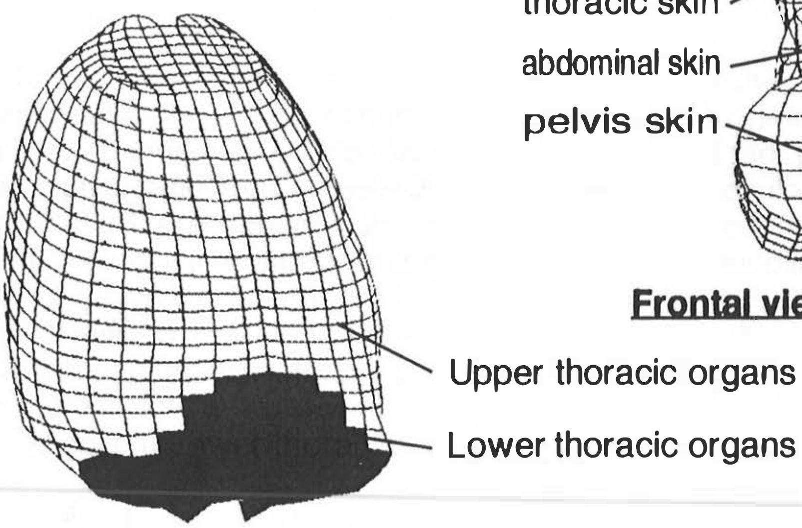
Frontal view of the rib cage **Lateral view of the rib cage**



Shoulder bones



Frontal view of the external surface



Internal organs

Figure 2 : **Views of the finite element model**

RESULTS

To adjust the mechanical characteristics of the model, four corridors have been used (cf. Table 1) :
- impactor load versus deflection from frontal impact on the thorax, config. thorax3
- impactor load versus deflection from frontal impact on the thorax, config. thorax4
- impactor load versus deflection from lateral impact on the thorax, config. thorax9
- impactor load versus deflection from oblique impact on the abdomen, config. abdomen3

The responses of the model are compared to all of the corridors of the database. Due to limited space, a selection of corridors has been made, certain results are presented in the form of tables.

The validation results are presented in the following pages, figure 3, 4, 9, 10 and 12. For each plot window, the model curve is differentiated from corridors curves by black discks. The units used are : impactor or belt load (N), time (ms) and, deflection and displacement (mm). Exemples of simulation views are presented figure 5, 6, 7 and 11.

FRONTAL THORAX IMPACTOR TESTS - For every impactor test simulations, the impactor head is rigid and guided along the impact axis and is launched with an initial velocity; the mass is that of the tests. The boundary conditions, as well as the initial position of the subjects, are respected.

There are numerous frontal test configurations for the thorax which introduce dispersions originating from the performance, analysis and boundary conditions of the tests. However, the responses of the model are well correlated to experimental results for all the configurations. (figure 3 and 5)

LATERAL AND OBLIQUE THORAX IMPACTOR TESTS - In both lateral and oblique impact directions, the responses of the model are very close to those of the subjects tested. The range of energy is large and is representative of the one found in vehicle impacts. The model may be considered as reliable. Both in frontal and lateral, the deformation of the spine during the impact is analysed and verified. (figure 4 and 6)

BELT COMPRESSION TESTS ON THE THORAX - The difficulty of these simulations is to represent correctly the contact between the back and the horizontal rigid flat table, knowing that the curve of the spinal column corresponds to a sitting position. The chosen solution is an interface between the back and a rigid surface with a form close to the one of the back. The belt model and it's mechanical behaviour law were validated with specific tests performed on pieces of the belts used in the cadaver experiments (Song 1993). The belt is dynamically loaded by imposing a measured load at the two extremities of the belt. The deflections are measured at nine points on the thorax and compared to the validation corridors.

Table 3 : Results of configurations A and B - Deflections in millimeters at the nine points of measurement and comparison with the minimum and maximum values of the tests

	target	**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**10**
A	**Model**	16	15	9	3	4	6	17	13	6
A	**Min**	18	17	9	2	2	11	10	22	2
A	**Max**	26	32	19	11	14	16	26	40	11
B	**Model**	64	55	46	12	19	28	65	50	18
B	**Min**	58	40	15	0	8	19	38	59	12
B	**Max**	70	77	48	18	13	45	74	94	28

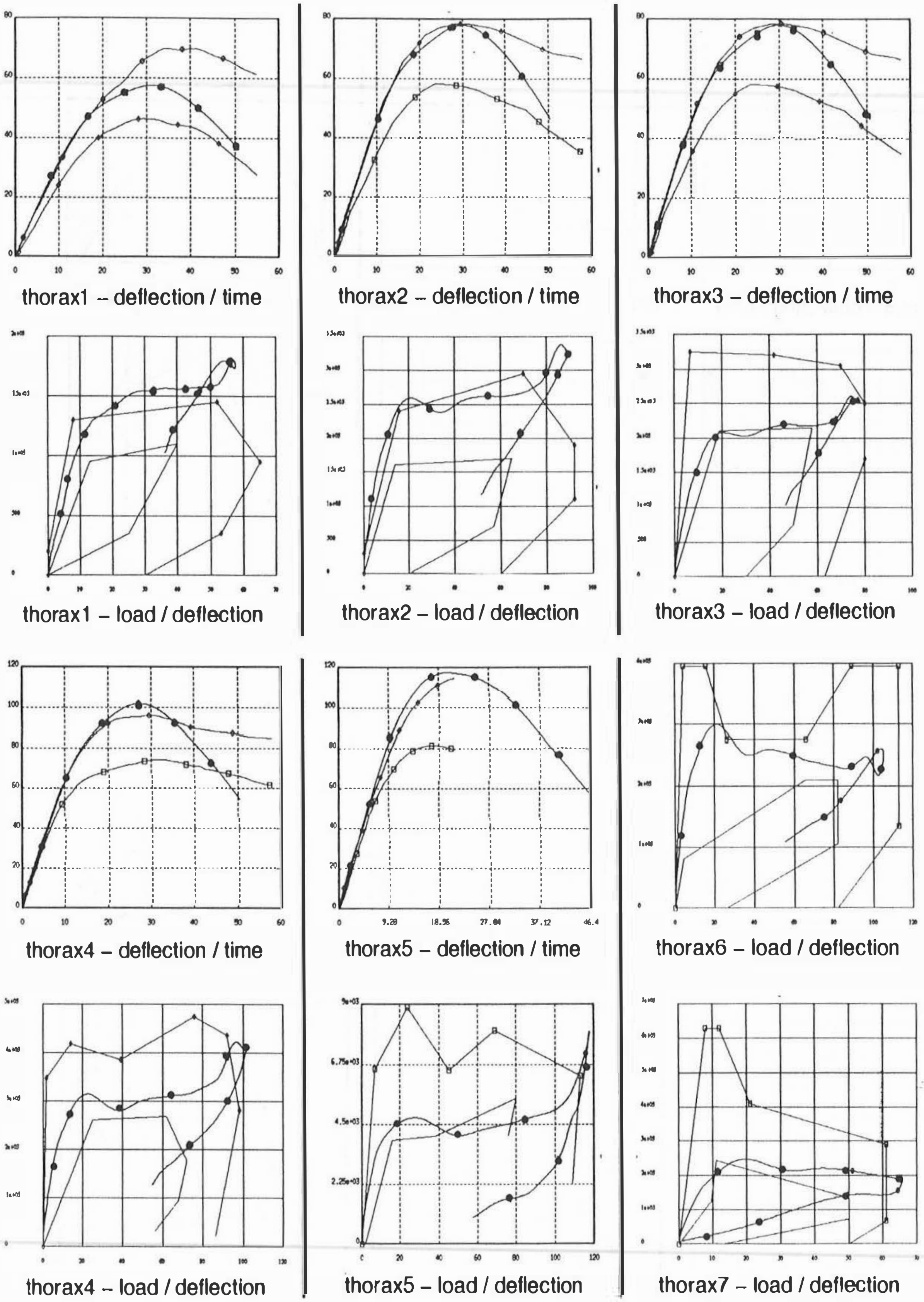


Figure 3 : **Frontal impact on thorax, configurations 1, 2, 3, 4, 5, 6 and 7 (cf. table 1)**

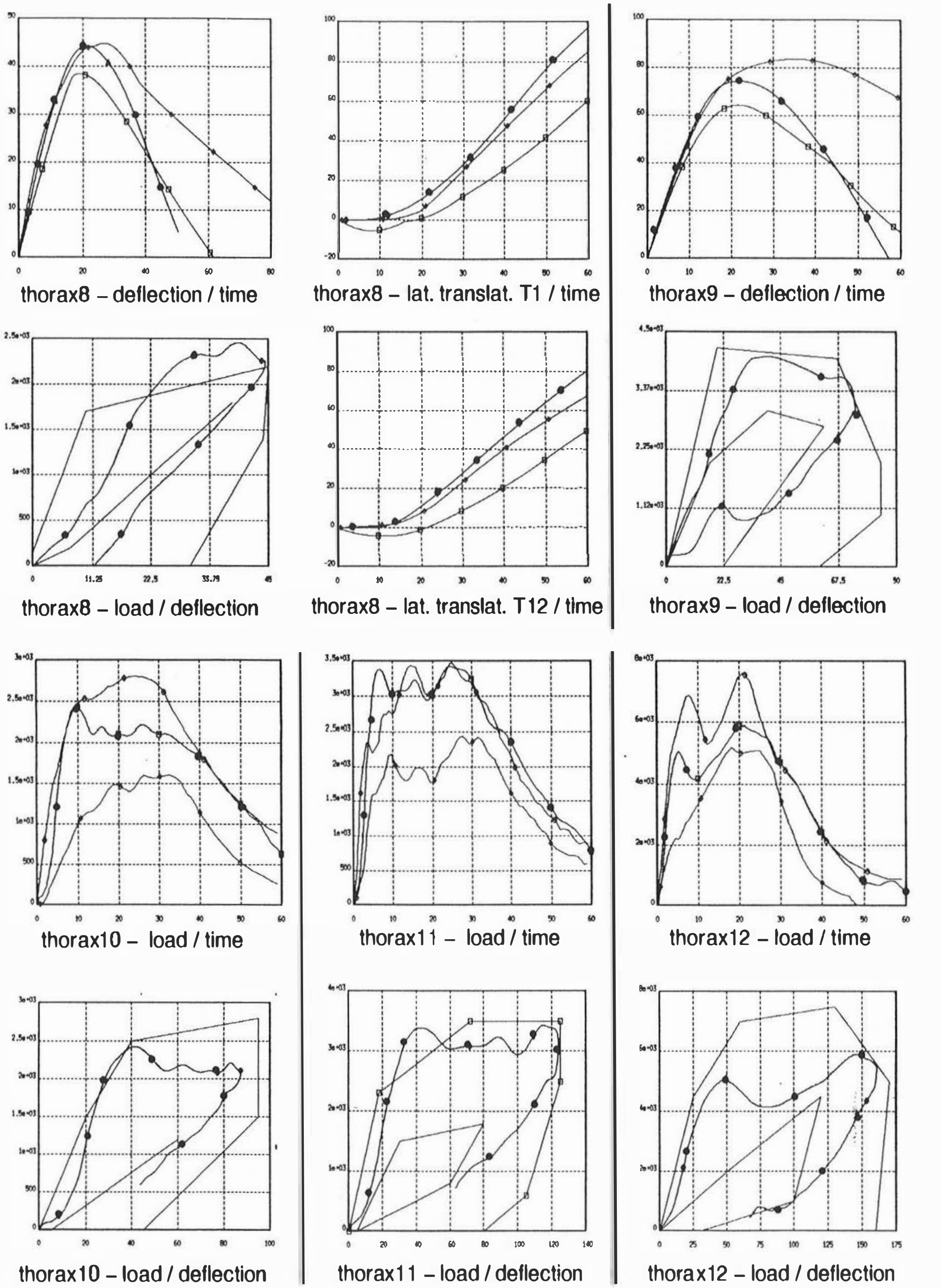


Figure 4 : **Lateral and oblique impact on thorax, configurations 8, 9, 10, 11 and 12 (cf. Table 1)**

Figure 5 : Frontal impact simulation on the thorax, configuration 2, time 0, 20 and 40ms

Figure 6 : Lateral impact simulation on the thorax, configuration 9, time 0, 20 and 40ms

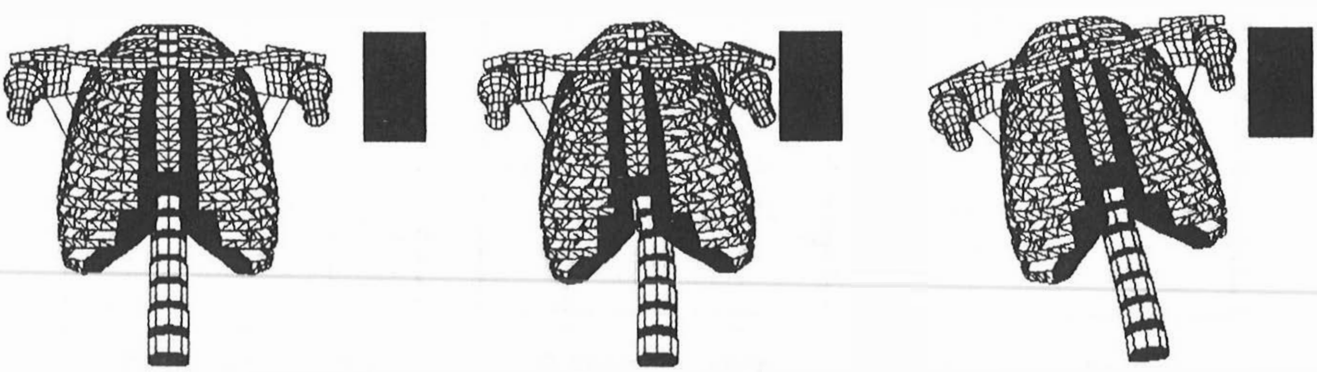
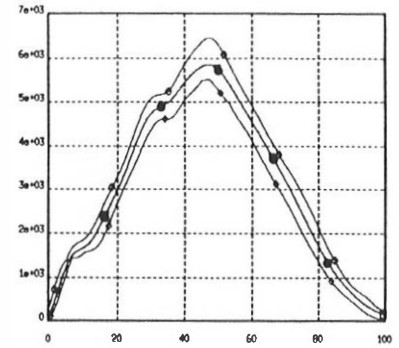


Figure 7 : Lateral impact simulation on the shoulder, configuration 1, time 0, 20 and 40ms



total load imposed to the two extremities of the belt

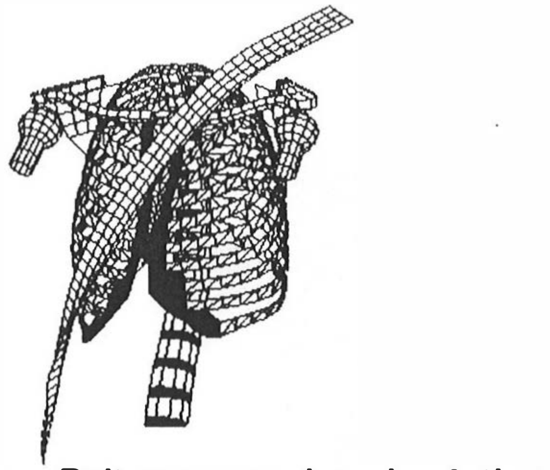
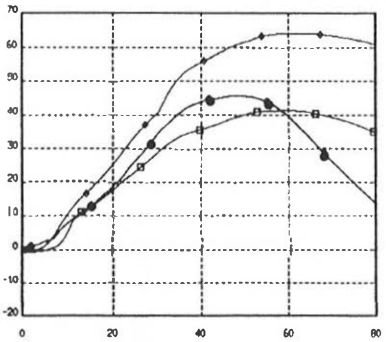
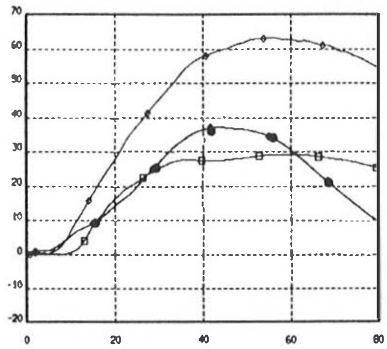


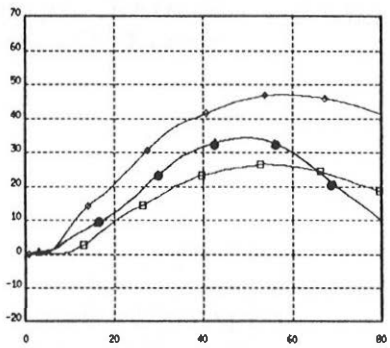
Figure 8 : **Belt compression simulation**



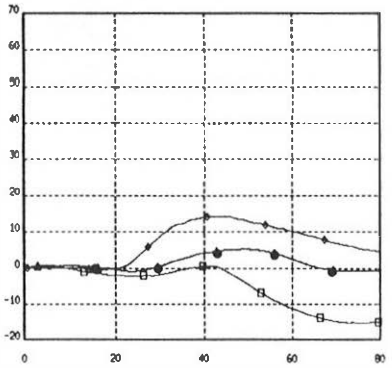
deflection point 1



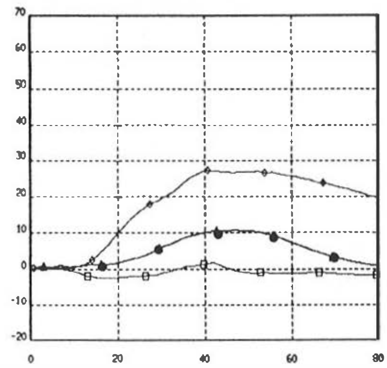
deflection point 2



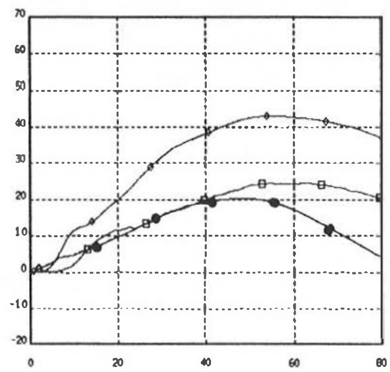
deflection point 3



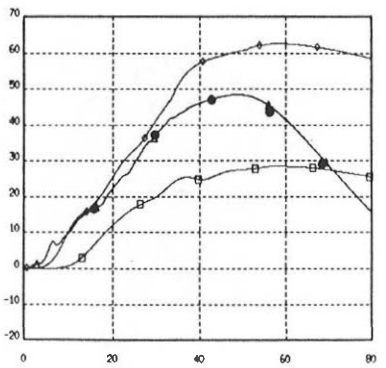
deflection point 4



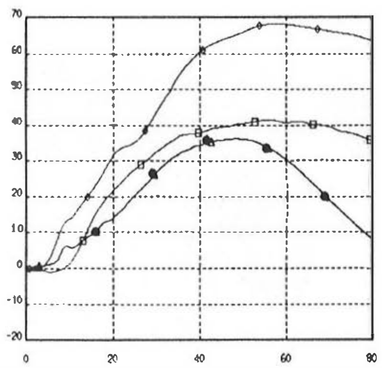
deflection point 5



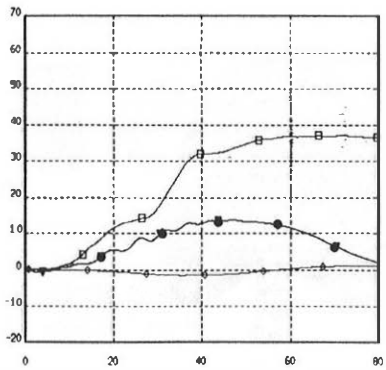
deflection point 6



deflection point 7



deflection point 8



deflection point 10

Figure 9 : **Belt compression tests on thorax, configuration C**

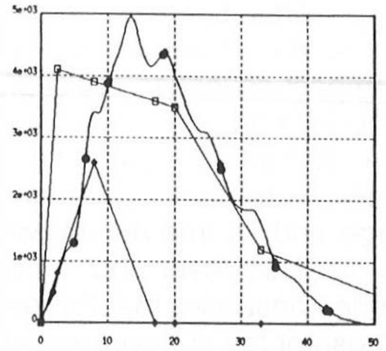


Figure 10 : **Lateral impact on shoulder configuration 1 (cf. Table1)**

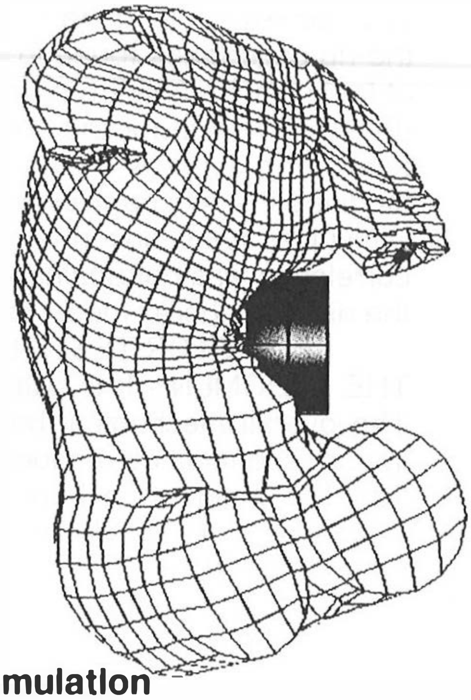


Figure 11 : **Oblique impact on abdomen simulation**

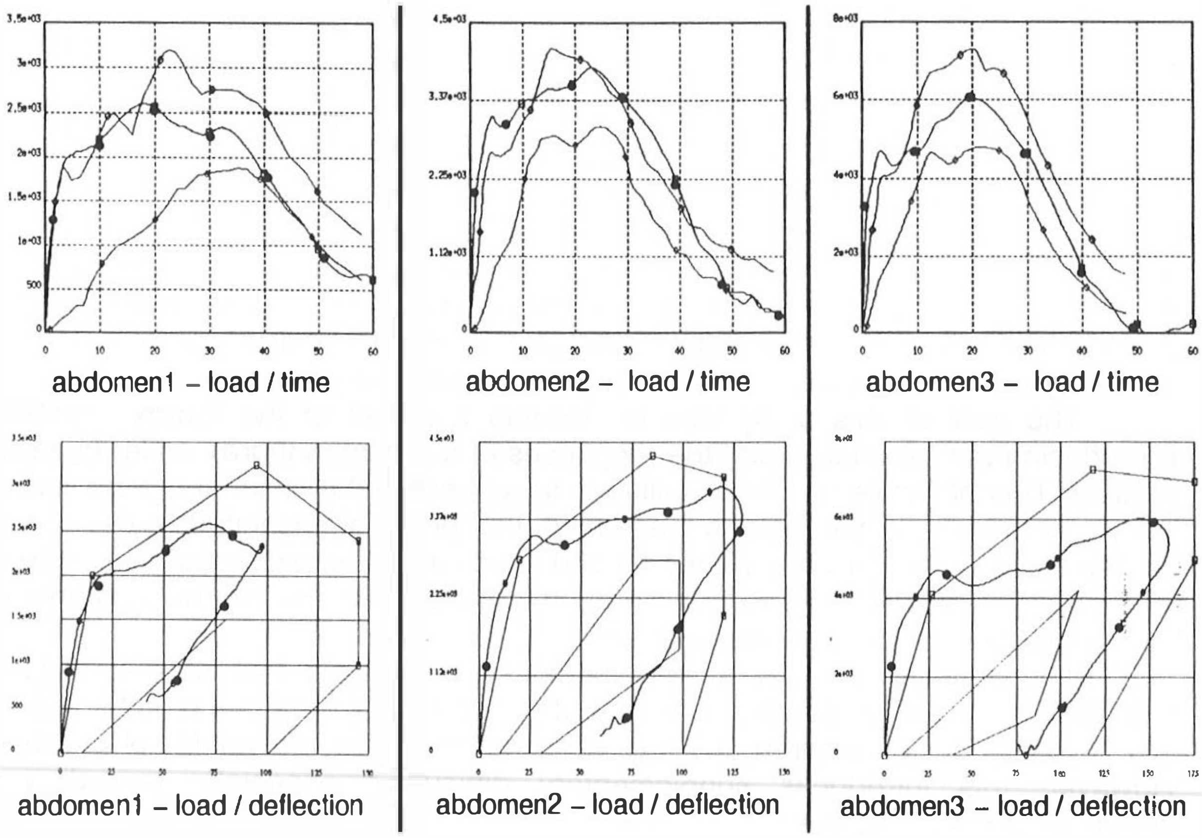


Figure 12 : **Oblique impact on abdomen, configuration 1, 2 and 3 (cf. Table1)**

The results of the loading configuration C are presented in figure 9.
The belt experiments have the advantage of their load configurations being close to the rib-cage loads found in real accidents. The impact severities are representative of fairly low severity frontal car accidents, even though rib fractures occur in the tests (five on average for the configurations B and C). This point is probably due to the age of the subjects tested. The fact that the back rests on a rigid plane probably modifies the mechanical behaviour of the ribcage and its interaction with the belt. Difficulties were encountered in the simulation of these tests and a high level of correlation was not attainable. Nevertheless, we demonstrated that the responses of the different thorax model targets are similar to those of the cadaver tested.

IMPACTOR TESTS LATERALLY ON THE SHOULDER AND OBLIQUELY ON THE ABDOMEN - It is difficult to find enough tests to validate the shoulder model. The overall mechanical behaviour under load is nevertheless acceptable and the impact scenario corresponds to the descriptions from Meyer (1994) (figure 7 and 10). The oblique mechanical behaviour of the abdomen is of good quality for the three levels of impact energy, for this segment the model is considered as reliable. (figure 11 and 12)

DISCUSSION

Quantitatively, the test data and the model responses compare well. These comparisons are performed in terms of the physical magnitudes describing the kinematics and the global deformations. Qualitatively, many verifications have been carried out with respect to the films or the deformation of chest bands. However, the validation base is not yet sufficiently comprehensive. For instance, many of the tests taken into consideration have been carried out with an impactor of a mass close to 23.4kg and the shoulder needs further validation. Verification of the model's responses by comparison with lateral impact sled tests (Irwin 1993) would be desirable, however, to achieve this, deformable pelvis and limb models must be added. Further progress must be made in defining the characteristics of the materials used. New experiments are indispensable particularly to characterise the dynamical behaviour of the soft tissues and to precisely distinguish in terms of geometry and of stiffness the different regions of the internal viscera.

The goal of this study was to develop a model of the thorax capable of reproducing the kinematics and the responses of the human thorax under dynamical loading. This objective has been fullfilled. In respect to the lesional criteria, there is good correlation of the criteria based on the physical magnitudes used in the comparisons made between the model and cadavers i.e. impact load, deflection and VC. The TTI and the thorax acceleration criteria, because of the contreversies persisting about them, were not analysed.

To compare to other similar models, for example that of Planck (1994) validated in the frontal direction and that of Huang (1994) in lateral direction, we can say that not having favoured any one direction reinforces the validity of our model. However, this introduces supplementary difficulties especially regarding the shoulders and the ribcage.

Given the database of the validation tests, the model is representative of the behaviour of a person aged approximately 60 years. It should be noted that the cadavers do not have any muscular action, so the mechanical behaviour predicted by the model is that of an occupant completely surprised by the impact. It is therefore unfavourable compared to many real-life cases.

The model is already in use for the definition of test protocols on cadavers with airbags. The tool has allowed us to define the test conditions and the level of severity required to conduct the experiments at levels close to lesional thresholds.

There are many facets to the use of this model : optimisation of the seatbelt/airbag couple, studying the influence of seatbelt load limiters, designing the door to improve lateral impact protection, etc..

CONCLUSION

A test database has been constituted. A model of the thorax and shoulders of the 50^{th} percentile adult male occupant in the driving position has been developed. Its behaviour in frontal, lateral and oblique impact has been evaluated quantitatively and qualitatively, it is coherent with respect to the cadaver tests used. A single model is thus capable of predicting the kinematic and responses of the human being in all types of automobile impact, with good precision. Work remains to be done to increase the reliability of certain parts of the model. Nevertheless, this model is already being used in a wide range of applications in addition to models of the standardised dummies. This study has demonstrated the reliability of the construction of such a model. The method used is being extended to the whole of the human body with the prospect of obtaining a predictive model for lesional risk in the automobile impact environment.

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