Finite-element human body model for automotive safety

From a kinematic model towards an injury model

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ABSTRACT. *This article presents a finite element human body model, used in the Laboratoire d'Accidentologie, de Biomécanique et d'études du comportement humain of French car manufacturers for its occupant safety research programs. The working-out, the main construction steps of and the validation process are described. An example of application, which consists of a comparative study in frontal crash with a dummy model, illustrates the interest and limitations, through the most significant results. After a basic study, applied to the development of a new thoracic criterion highlighting real safety, the approach to improve the model into a 3D breakable model is shown. The bone condition differences are taken into account. Several future research prospects are exhibited in order to reinforce the human body model.*

RÉSUMÉ. *Cet article présente un modèle éléments finis de l'être humain, utilisé au laboratoire d'accidentologie, de biomécanique et d'études du comportement humain des constructeurs français pour ses recherches sur la sécurité des occupants de véhicules automobiles. La genèse, les grandes étapes de construction et de validation du modèle sont décrites. Un exemple d'application, qui consiste en une étude comparative du comportement au choc avec un modèle de mannequin de choc frontal, illustre son intérêt et ses limites, avec l'appui des résultats les plus significatifs. Après une étude de principe, appliquée à la définition d'un critère thoracique mettant en valeur la sécurité réelle, la démarche de transformation en un modèle lésionnel omnidirectionnel est montrée, avec la prise en compte des différences interindividuelles. Quelques perspectives de recherches permettant de le consolider sont exposées.*

KEYWORDS: *finite-element model, human being, automotive safety, injuries.*

MOTS-CLÉS : *modèle éléments finis, être humain, sécurité automobile, blessures.*

1. Introduction

At the Laboratoire d’accidentologie, de biomécanique et d’études du comportement humain (LAB) of the french car manufacturers, modelling has been used for more than twenty years to study the loads undergone by an occupant in an automotive crash. The first models, built with rigid bodies and kinematic joints, reproduced the external loads, without describing however the human body deformations.

With the first applications of the finite element method for the car crash analysis, in the late 1980’s, it became possible to integrate some biomechanical knowledge in a more representative human body model, with help of accidentology and tests on post mortem human subjects (PMHS). Thus, after several years of collaboration between industrial, academic and associative partners, the finite element human body model used at LAB became a tool impossible to circumvent for the research undertaken on the occupant safety.

Working with the fast dynamics computer codes with large deformations in explicit method (Radioss), it consists of 12 000 shell, solid and spring elements, for a typical time step of 1.5 µseconde. Its application field covers most of the impact patterns met in automotive safety.

After an overview of the project history, issues and choices, the main development and kinematic validation steps are described. More than hundred corridors of biofidelity, based dynamic PMHS tests were used.

As an application example, a comparative study with a dummy model illustrates the interest of the model and its limitations. The most significant results are given.

Recent work was undertaken to reproduce skeleton fractures. After the presentation of a study of principle, leading to the development of a new thoracic criterion on dummy, an original step is presented to take into account the individual differences of osseous resistance.

From the current model limitations, showing that the knowledge on materials composing the human body has to be improved, some prospects for research which would make possible to reinforce it are exhibited.

2. Finite-element human body model

2.1. *History and issues*

At the end of the years 1980, the evaluation of the passive safety offered to the occupants of the motor vehicles passed already by the use of anthropomorphic dummies in car crashes representing the main accidents observed on the road. However, the accident studies, relayed by the experiments on PMHS, raised the maladjustment of the dummies and the associated criteria in certain cases. As an example, the Hybrid III dummy, used for frontal impact, was developed in the United States to evaluate the protection of the not-belted driver.

In the same time, the numerical tools were developed to simulate the vehicle deformations. Their use in the design process made it possible to improve the mechanical metal assembly behaviour in case of impact, in addition to the full-scale tests. The development of numerical dummy models permits to optimize the occupant restraints for regulatory specifications.

With the will to have a numerical tool representing a better biofidelic human being subjected to different automotive impacts, in order to add a relay with the accident surveys to reach out for a more real safety, the LAB was asked to develop a numerical human body model as following:

– feasibility study (years 1991-1992), with the assistance of the authorities, and in collaboration with the INRETS, the ENSAM and the university of Lyon,

– development (years 1993-1997) of a finite-element 3D-directional model, in collaboration with the ENSAM,

– first uses and improvements, with the CEESAR, since 1998.

2.2. *Schedule of conditions*

In its final version, the project declined the following requirements:

– finite-element 50th percentile male model in driver position,

– model validated in frontal, side, oblique and rear impact,

– emphasis on the neck, the thorax, the abdomen, the pelvis and the spine with deformable elements, the other parts being defined in rigid bodies,

– compatibility with the car models, regarding the size with a maximum of 10 000 elements, and a time step greater than 1 µs,

– numerical validation in an automotive environment, in frontal, side and rear impacts.

2.3. *Implementation*

2.3.1. *Computer code*

The choice of the computer code was imposed de facto at the origin of the project, answering the criterion of compatibility between the manufacturers, and the need for treating in dynamics the great deformations with the means of calculation placed at the disposal. Thus, it was used an explicit code (Radioss), with a typical constant time step of 1.5 µs, despite of a mass increase during calculation, kept lower than 1%.

This type of code is usually dedicated to the metal structure analysis, comparable to thin shell assemblies. In our case, solid elements constitute most of the model mass. A feasibility study showed that the code could integrate visco-elastic materials with low propagation velocity, taking into account cautiously the element stability.

2.3.2. *Geometry and mesh construction*

The data which made it possible to generate the geometry are the following:

– model envelope starting from geometrical data of a 50th percentile male occupant, in driver position (Robbins *et al.*, 1983),

–scanner sections of a 50th percentile male subject in reclining position for the rib cage and the shoulders,

– dimensions of the cervical discs and vertebrae (Maurel *et al.*, 1993, Panjabi *et al.*, 1993),

– spine curvature, and rib inclination (Dansereau *et al.*, 1988, Rebiffé, 1982, Martens),

– definition of the pelvic geometry (Reynolds *et al.*, 1981) and orientation (Césari *et al.*, 1994),

– envelopes of the upper and lower limbs, as well as the head, from the Hybrid III dummy.

With the tools used by the engineering and design departments of the car industry, the mesh was obtained by using the extrusion method, taking into account the complexity of the pelvis especially, the concern of minimizing the number of internal interfaces and insuring element size homogeneity. Figure 1 shows the final outer model envelop after assembling.

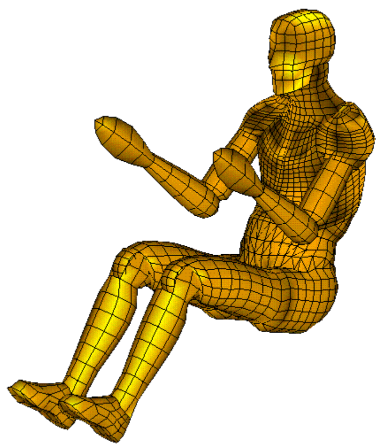

Figure 1. *Oblique view of the external envelope of the human body model*

2.4. *Description of the model components*

The model components were described by Lizée *et al.*, in 1998. The main elements allowing the whole model understanding are reported. The evolutions carried out since 2 000 are also mentioned.

2.4.1. *Neck*

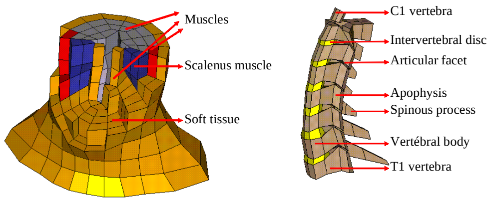

Figure 2. *Neck components*

The neck spine is made of rigid vertebrae, and deformable discs with an interface declared between articular facets (figure 2). Solid elements define differentiated muscles. Springs with visco-elastic properties represent ligaments, as well as the C1 vertebra-occiput and C1 vertebra-odontoid connections.

2.4.2. *Thoracic and lumbar spines*

Solid elements compose the thoracic and lumbar spines. The vertebral units are simplified, with deformable brick disc elements, whose properties are tuned to insure equivalent joints.

2.4.3. *Internal organs*

The internal organs are undifferentiated. A distinction is however made between the thoracic organs and the abdomen, in term of mechanical properties. The mesh is continuous and the nodes coincide with those of the rib cage and the spinal cord.

Maxwell models are used for visco-elastic materials. Thus a modulus of compressibility K and a function of relaxation for shear are defined:

$$G(t) = GI + (G0 - GI) * \exp(-\beta t) \text{ and } K = E/(3.(1-2\nu))$$

with: G0 = short time shear modulus, GI = long time shear modulus

Young’s modulus (E) and, Poisson’s ratio(ν).

β = damping ratio.

This approach is also used for muscles, although the literature shows their strong nonlinearity.

The use of other mechanical laws taking of account the deformation speed was tried, with however unacceptable computing times.

2.4.4. *Shoulder joints*

By its large mobility, and its important implication as well in side impact as in frontal impact, shoulder modelling was tricky to implement. After various attempts, it was decided to model the clavicle, scapula and humeral head without ensuring the mesh continuity with the surrounding muscles (figure 3). Only the clavicle is deformable. Generalized springs define all joints. Contact clavicle-rib and clavicle-skin interfaces with the penalty method are declared.

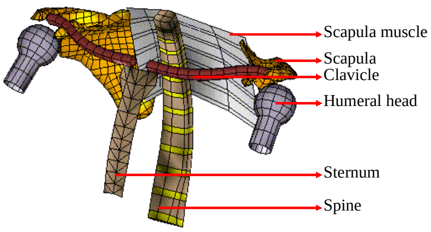

Figure 3. *Modelling of the shoulder*

2.4.5. *Rib cage*

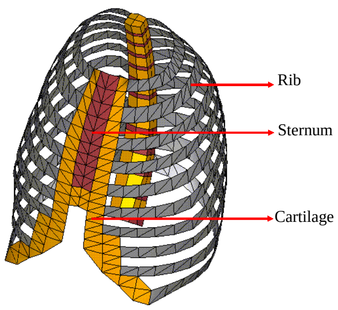

Figure 4. *Rib cage model*

Triangular shell elements constitute sternum, ribs and cartilage. This simplified approach defines a way to limit propagation of hourglass modes, keeping in mind that the goal was to develop a model and validate its kinematics response. In the original version, costal-vertebral joints are modelled by a direct connection from the rib extremity to the edge of the corresponding vertebra.

2.4.6. *Pelvis and femoral heads*

The pelvic skeleton is built with elements shells, taking into account the sacrum-iliac joints and the symphysis pubica (figure 5). Moreover, regarding the stiffness, the contribution of the trabecular bone was neglected in front of that of the cortical bone (Dalstra *et al*., 1992). The acetabulum and the great trochanter are declared in rigid bodies and connected by a generalized spring forming the hip joint.

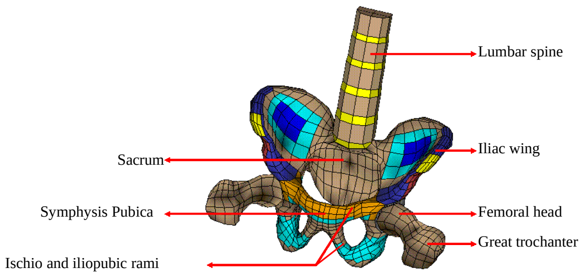

Figure 5. *Pelvis model*

2.4.7. *Skin*

On the outer envelop of the solid elements constituting the neck, shoulder, thorax and pelvis muscles, shell elements define skin with an elastic material law.

2.4.8. *Upper and lower limbs, head*

All of these elements are declared as rigid bodies. However, depending of the purpose, adaptations were realized, such implementation of deformable arms for the studies in side impact (figure 6).

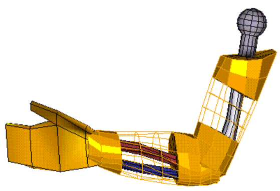

Figure 6. *Deformable arm model*

2.5. *Validation*

The validation of the model required an in-depth analysis of all the tests carried out on non-embalmed SHPM, available in-house or published in the literature. The selection criteria take into account the control of the boundary conditions, the inter-individual differences in term of anthropometric characteristics and masses, and finally available measurements (basically efforts and displacements). More than one hundred corridors known as "biofidelity references" were built (Lizée *et al.*, 1998) as data bases of model validation in dynamics. The types of tests are as follows:

– sled tests (frontal, side and back impact),

– pendulum tests with various masses and speeds, on the thorax, abdomen, pelvis and shoulder,

– thoracic compression with belt restraint,

A validation of segment masses, centers of gravity and inertias was also carried out.

The model reproduces the kinematics of the different segments with an acceptable level. Its mechanical behaviour, judged in term of effort-displacement curves is overall satisfactory.

Apart from the tests conditions used for the validation, the model was tested in violent configurations, in order to test its robustness. The model is numerically stable and the hourglass energy remains lower than 10% of the internal deformation energy in the most severe cases.

2.6. *Conclusion*

Using available data for the geometry and the response, a 50^{th} percentile male human body model was developed using the finite element method. The model meets the specifications. It defines a simple and robust tool allowing the investigation of various restraint systems and the deformations undergone in automobile crashes by a human being (figure 7).

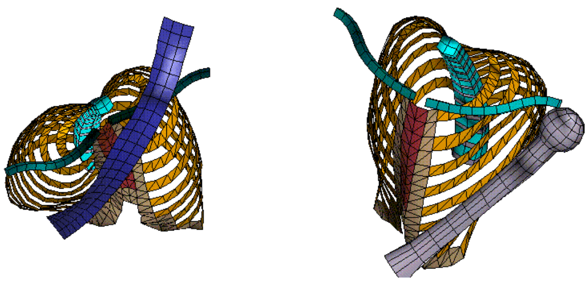

Figure 7. *Rib cage loaded by a seatbelt (on the left) and a door in side impact (on the right)*

The human body model is able to reproduce submarining (Leung *et al.*, 1982), *i.e.* the slip of the pelvic belt above the iliac crests (figure 8).

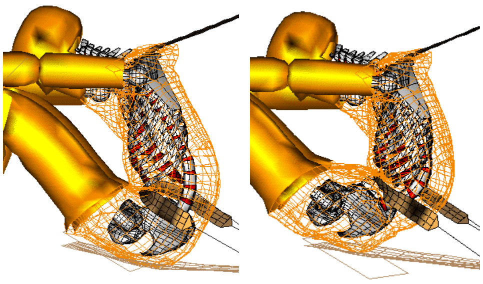

Figure 8. *Illustration of submarining phenomenon*

3. Example of use: compared analysis of the human body model and a Hybrid III dummy model in frontal impact

In order to quantify the difference between the results obtained with the human being and those obtained with the dummy, a study was carried out in frontal impact (Baudrit *et al.*, 1999), using the human body model and the Hybrid III 50th percentile dummy developed in the BRITE-DUMOCS European project.

3.1. *Environment*

The selected configuration represents a severe frontal impact, with a speed of 56 km/h for a 630 mm crushing distance. The whole driver occupant restraint systems are modelled.

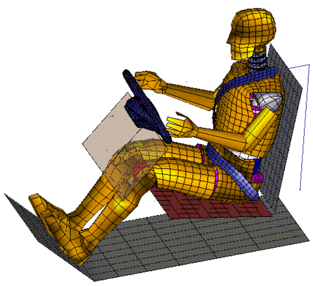

Figure 9. *Overall view of the environment with dummy model*

The variables of the study related to the load limiter design features, the presence or not of the inflatable bag and its characteristics. Some significant results illustrating the interest of the human body model are presented hereafter.

3.2. *Examples of results*

3.2.1. *Belt load comparison*

Figure 10 compares the outer pelvic belt loads. A significant difference is noted. The dummy model induces a more important and earlier restraint load, with a 1500 N difference for the selected configuration.

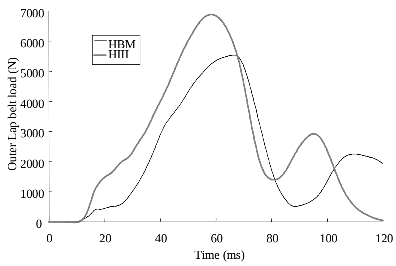

Figure 10. *Comparison of outer lap belt loads*

The upper shoulder load duration is higher for the human body model (figure 11), inducing a longer load limitation belt length.

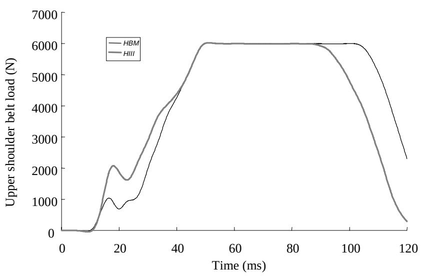

Figure 11. *Comparison of upper shoulder belt loads*

The two occupant masses being the same, the explanatory assumptions are:

– difference of lumbar spine stiffness,

– difference of shoulder and "flesh" in front of the pelvic skeleton stiffnesses,

– chest shape difference.

Note that several evidences come to reinforce these observations. The recent accidentology studies show that with the same impact violence, the lengths of load limitation, measured under tests with HIII dummy are lower than those observed in real-world accidents.

In addition, the lap belt loads observed differ between HIII dummy and SHPM, and between HIII and THOR dummies (Vezin *et al.*, 2002). The two dummy have the same segment masses, however the Thor dummy has a more biofidelic lumbar spine.

3.2.2. *Study of the thoracic deflection for two restraints*

For the same impact, it is studied the effect of two restraint systems on the thoracic deflection, which are:

– 6 000 N load limiter device.

– 4 000 N load limiter device plus inflatable bag with variable characteristics.

These two devices represent two generations of thoracic restraints, providing European cars in recent years.

Note that the air bag restraints the head and the thorax at the same time.

Figure 12 shows the thoracic deflection, for the two restraint systems. The average value obtained for various restraint systems with inflatable bag is displayed, as well as the standard deviation.

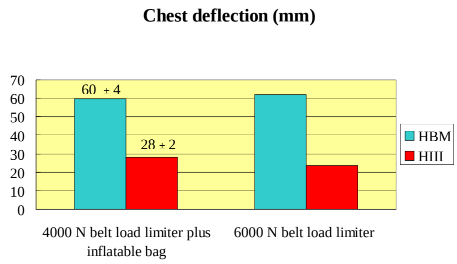

Figure 12. *Effect of various restraint systems on the thoracic deflection*

The thoracic deflections are comparable for the human body model for the two restraint systems studied, with a decrease of approximately 4% for restraint with air bag. For the HIII dummy model, the deflection is 20% higher on average with inflatable bag. With a slightly less marked tendency, tests carried out on HIII dummy show the same trends (Petitjean *et al.*, 2002).

These results are not in accordance with the accidentology results (Forêt-Bruno *et al.*, 2001) which show that the belt load explains the thoracic injuries. Two conclusions can be drawn:

– the HIII dummy is not appropriate to evaluate some restraint systems which show their effectiveness on the road,

– despite the use of a more biofidelic tool than the dummy, the results poorly correlated to the real world accident data. It suggests that the thoracic deflection, measured in the middle of the sternum, reflects in a very limited way the injury risk if two very different restraint systems are superimposed. Note that the belt restrains the thorax through concentrated loads while the air bag exerts a rather homogeneously distributed load.

From this report, several studies were undertaken in order to provide on the one hand a more discriminating dummy criterion, and, on the other hand, to transform the kinematic model into an injury level model, able to predict the risk of rib fractures in a realistic way.

4. Injury level model

4.1. *Basic study*

Taking into account the previous conclusions, it was necessary for the mathematical tool to reproduce the rib fracture mechanisms, whatever the load pattern.

The first evolution of the human body model consisted in improving the rib material law, with elastoplastic properties and limited strains, by analogy with the work completed on pelvis (Besnault *et al.*, 1998). Thus, equipped with a "microscopic" criterion, the thorax of the human body model can "sustain injuries". The validation is a complex task. It requires a precise documentation of the SHPM tests used to develop the criterion. It includes the boundary conditions, the subjects characterization as a function of age (anthropometry, geometry and individual mechanical rib properties), and, at last, the injury assessment. Moreover, the loading patterns must be varied and be representative of the automobile crashes.

It was initially decided to undertake a basic study by considering only one occupant (fixed material law). The selected injury threshold corresponded to a number of rib shell elements destroyed at the end of calculation.

This model was used with several frontal thoracic automotive restraint systems responsible for concentrated belt loads only, distributed air bag loads only, or combined (belt and air bag) loads. The purpose was to assess the relevance of the thoracic deflection as an injury criterion on the human body model and to propose an alternative on Hybrid III dummy (Petitjean *et al.*, 2003). Some results are presented hereafter.

4.1.1. *Study of the relevance of the thoracic deflection as an injury criterion*

From the simulation results, the injury risk, translated by the appearance or not of 7 shell elements destroyed on the whole thorax, is calculated according to the mid-sternum deflection. Figure 13 shows that the threshold with a 50% risk depends on the load pattern. The deflection was 42 mm for a belt loading and 72 mm for a distributed loading. In comparison, the Hybrid III dummy gives the same tendencies with, for the same risk, 50 mm with belt and 61 mm with distributed loading (Mertz *et al.*, 1991, 1997). These two elements suggest the strong dependence of the loading pattern on the risk established from the deflection.

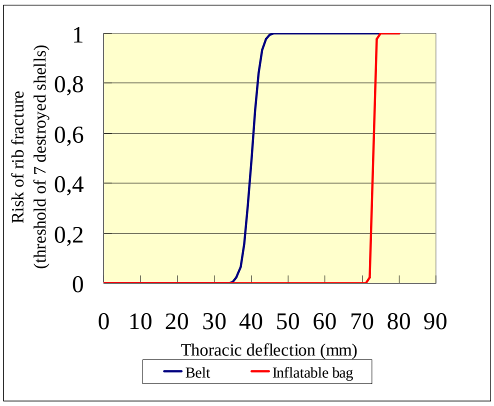

Figure 13. *Injury risk on the model according to the chest deflection*

4.1.2. *Proposal for a new criterion of thoracic injuries, the EDC (equivalent deflection criterion)*

	Current study		Accident database
	Deflection criterion	Equivalent Deflection criterion	Belt load criterion
4 kN belt load limiter plus inflatable bag	27 %	1 %	1 %
6 kN belt load limiter	13 %	11 %	18 %

Figure 14. *Risk of severe lesion for various injury criteria*

Based on the information available, a criterion using the thoracic deflection and the shoulder belt load was proposed (Petitjean *et al.*, 2003). An equivalent thoracic deflection is calculated from two quoted measurements, the idea being to estimate the belt and air bag contribution in the total deflection. This criterion was established on its principle from the human body model and was transposed on the Hybrid III dummy. It gives consistent results in term of risk compared to road reality (figure 14). In addition, it shows the major contribution of the belt on the injury risk, in the case of combined restrained systems. Its field of application is dedicated to the restraint systems used under normal conditions. Out of Position situations were not investigated.

4.2. *Reinforcement work in progress*

The generalization of the active systems aboard vehicles led the LAB to dedicate researches to the occupant safety in case of Out-Of-Position (OOP). A new need for understanding of the specific physical phenomena involved in this type of active restraint systems (inflatable devices such as airbag or cushion, pretensioners…) appeared. The need of adequate modelling (Petit *et al.*, 2003) arose next. Moreover, the evaluation of the criteria relevance on dummy remains a challenge. Indeed, the tests on SHPM, necessary for the understanding of the phenomena are limited in number. The development of injury curves, with censored data from small biased samples, remains a major issue, while the ambition is to protect the whole population.

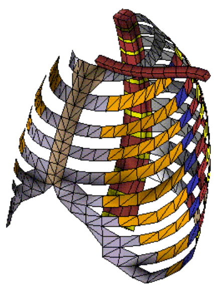

Figure 15. *Development of the rib cage*

The human body model was modified in order to improve its ability to be used as a complement to the cadaver tests.

The rib cage geometry was modified (figure 15) and the material properties of soft tissues were adjusted. The rib properties were modified too as following:

– different inertial and material properties were set according to their curvilinear abscissa (Chen *et al.*, 1970),

– the mechanical rib properties were defined based on SHPM characterization (Got *et al.*, Sacreste *et al.*, 1982).

The injury level thorax model validation process is summarized on figure 16.

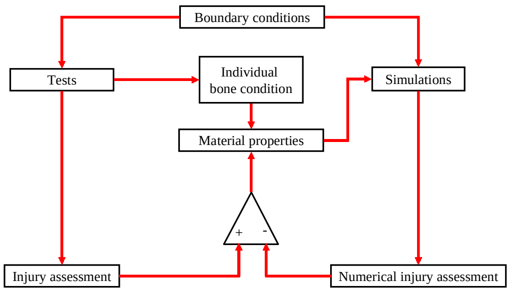

Figure 16. *Model validation process*

For each configuration, the results obtained using the model and real test injury assessments are compared regarding the fractured ribs. By an iterative process, the mechanical properties are adjusted in order to minimize the difference between the numerical and real injury assessments. The relationship between the mechanical properties and the characterization of the subjects is established and analyzed.

The distribution of the fractures observed on the rib cage was also analyzed and the model was improved to remove the unrealistic fractures observed at the beginning of the study.

The selected configurations cover most of the data placed at the disposal by the literature, as well as tests carried out with the CEESAR on SHPM. The various load patterns are represented (belt, inflatable bag, impactor, rigid wall…), as well as the various impact directions (frontal, side, oblique).

5. Discussion

As shown in the first part, the mesh of the human body model is not refined and simplifications have been chosen in order to obtain a tool with an acceptable time step. As an example, the right rib geometry is not represented, requiring calculation of the equivalent thicknesses and material properties in order to comply with bending stiffness. Despite these short cuts, the first results are promising, with the existence of a relationship between the mechanical parameters and the characterization of the subjects. They will have to be analyzed by considering that the differences in anthropometry were not taken into account, and that, by assumption, the data are consistent.

Note that this reasoning is applicable whatever the model complexity, the consistency of the final results being a tangible comparative reply. For the pelvis, the same step is in the application process, based on subsystem tests where the geometry was precisely documented (Guillemot *et al.*, 1997).

The applications of the human body model are varied. They mainly concern the research for new protection technologies and new criteria on dummies. In addition, the model makes it possible to refine test protocols on SHPM, according to the needs.

The tests on SHPM are better and better documented. The instrumentation of the skeleton by gauges provides useful information on the bone deformations and times of rib fracture occurrence which appears interesting to exploit, in connection with the sub-system tests of bone resistance characterization. The systematic individualization of the model seems to be another useful research orientation for a more accurate definition of the injury risk.

The whole research focussed on the bone structures. For the soft organs and tissues, knowledge remains very incomplete, both on the mechanical response and on the injury mechanisms. The implementation of tests reproducing the injuries observed on the road is more complex, because it requires the control of the SHPM reconditioning. Moreover, internal measurements, except the pressure, still have to be developed. As of today, the modelling of the phenomena must be assumed to be correct, even though simplified, and able to bring realistic answers. Some samples of tests with known boundary conditions exist in literature (Trosseille *et al.*, 2002). They will be exploited as much as possible.

6. Conclusion

A 50th percentile male human body model was developed, using the finite element method, after several years of collaboration between industrial, university and associative partners. Validated in many impact configurations, it is a simple but robust tool which does not require heavy calculation with a limited number of elements.

In its original version, its use made it possible to obtain a critical glance on the biofidelity and protection criterion of the regulatory dummy. In order to improve the relevance of the results, the model is on the way to become an injury level model for the thorax and the pelvis, by considering the differences in osseous resistance. Thus, it synthesizes a whole of knowledge resulting as much from biomechanics than accidentology. It is suitable for improvements of understanding of the injury mechanisms, and defines finally a tool for evaluation of real safety.

The focus was related to the skeleton. For the internal organs, the task of validation on the injury level seems more difficult, taking into account the limited knowledge of the injury mechanisms and the dynamic behaviour of constitutive materials. Note that even a simplified modelling can bring concrete answers with the assistance of controlled tests.

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