HUMAN RESPONSE TO A FRONTAL SLED DECELERATION

Philippe Vezin, Karine Bruyere-Garnier, François Bermond
INRETS - Institut National de Recherche sur les transports et leur Sécurité
LBMC - Laboratoire de Biomécanique et de Mécanique des Chocs
Bron, France

ABSTRACT

Sled tests were conducted with cadavers to obtain data on the behaviour of the human body under frontal deceleration. Two series of three tests were conducted corresponding to two levels of severity and types of restraint systems. The restraint systems used for the tests included a 4kN force-limited shoulder belt and a static lap belt. A driver airbag was mounted for the series at highest severity. The results show that the restraint systems and the test conditions modify the behaviour of the cadaver. The restrained conditions are not reflected in the same way for different segment of the body. The influence of the airbag is well observed for the chest spine behaviour, and not for the first thoracic vertebra. This study suggests that the chest acceleration (T8) could be a better predicting parameter than the T1 resultant in a combined parameters injury criterion.

KEY WORDS

BIOMECHANICS, CADAVERS, FRONTAL IMPACTS, RESTRAINT SYSTEMS, SLED TEST.

AT A EUROPEAN LEVEL, the assessment of the protection to a restrained occupant offered by a vehicle in case of frontal collision, is based on biomechanical data measured on the Hybrid-III crash test dummy described by the European Frontal Directive (October 1998). This dummy was developed in seventies and is based on biomechanical knowledge from that time. Although the dummy has been changed since then, it is not sufficiently able to assess the protection offered to the vehicle occupant and does not reflect current biomechanical knowledge (c.f. **ADRIA** European Project, Advanced crash Dummy Research for Injury Assessment in frontal test conditions, 2000). Within the **FID** European research program (Improved Frontal Impact Protection Through a World Frontal Impact Dummy), a dummy prototype, based on the THOR-alpha, will be developed together with a set of requirements for frontal dummies.

In the **FID** project, biomechanical tests were performed to fill the gaps in the biomechanical knowledge on human body behaviour during frontal impacts. This biomechanical data completed with a bibliographic study allowed to propose a set of response requirements to assess the biofidelity of an anthropomorphic Frontal Impact Dummy to the European Enhanced Vehicle-safety Committee (van Don et al., 2002).

Many cadaver tests have been done with different restraint conditions and subjects seated in the driver position (Kuppa and Eppinger, 1998, Kallieris et al. 1995). Generally these data are not easily comparable, because the instrumentation and impacts conditions (velocity, deceleration) were very different. The comparison with the dummies is also difficult. There are only few studies with whole and detailed similar measurements on the cadavers and on the dummies, carried out with the same tests and impacts conditions. One of the consequences is the introduction of a bias into the injury criterion that is intended for use with a dummy (Kent et al., 2001). For instance, the behaviour of the Hybrid III chest is humanlike at and above 4.6 m/s impact velocity but it may be stiffer than the human chest at lower impact conditions (Horsch and Schneider, 1988). Sled tests using Hybrid III

with belt or airbag suggested that the chest would behave stiffer than human chest. As a result the injury prediction based on chest deformation is often underestimated.

The objective of the present study (part of the **FID** project) was to obtain new data on the behaviour of the human thorax/shoulder complex under different frontal impact conditions. The same tests with exactly the same instrumentation mounted at the same location was also performed with standard Hybrid III and with the new THOR-alpha (Test device for Human Occupant Restraint) dummies to asses the biofidelity of these dummy with respect to the data presented here (Vezin et al., 2002).

This paper presents the information on the response of the human body in different frontal sled test conditions. The methods used for the tests are presented, an injury description is analysed. The accelerations of the thorax, the spine, the head and the shoulder are given and discussed.

METHOD

EQUIPMENT

Test conditions – Two series of three sled tests with Post Mortem Human subject – PMHS – were carried out. The first series – FID11 to FID13 – was conducted at 50 kph with a sled deceleration law close to the ECE R44-03 regulation (Child's restraint regulation), corresponding to a maximum deceleration of 22 Gs, and seat belt and airbag as restraint system. The second series (FID14 to FID16) was performed at lower velocity – 30 kph – and lower deceleration, about 15 Gs, close to the deceleration law used at the University of Heidelberg (Kallieris, 2001) and only seat belt as restraint system. Fig. 1 and Table 1 illustrate the different sled tests configurations.

Table 1. – Test conditions (FL: Force Limiter, AB: Airbag)

Test	Sled Velocity (kph)	Max. Sled Decel. (G)	Stopping Distance (mm)	Restraint system
FID11	49.26	-22.67	590	4kN FL + AB
FID12	49.78	-23.17	590	4kN FL + AB
FID13	48.49	-21.77	570	4kN FL + AB
FID14	29.90	-13.73	265	4kN FL
FID15	29.70	-13.73	270	4kN FL
FID16	30.06	-14.46	280	4kN FL

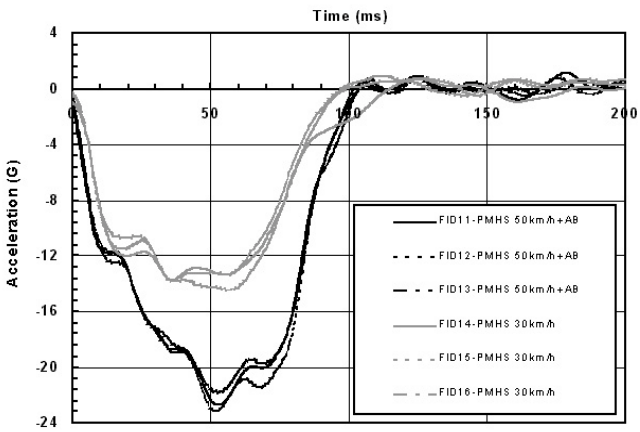

Fig. 1 – Sled deceleration time history.

Restraint system – Separate shoulder and lap belts restrained the subjects. The shoulder belt was equipped with a 4 kN force-limiting system. A torsion bar placed inside the retractor controlled the load in the shoulder belt by absorbing the energy through his deformation. The belt restraint was a standard production retractor system without pre-tensioning device. The pre-tension was made manually before the crash and was not recorded, for this purpose the retractor of the shoulder belt was blocked prior to testing. The shoulder belt ran over the left clavicle. The loads in the belts during the impact were measured.

For the series at 50 kph a standard airbag was mounted on a steering wheel fixed on the sled. The airbag was triggered electronically 20 ms after the impact. The shoulder seat belt and the airbag came

from the same restraint system of a French common car. The hands were strapped on the steering wheel in a natural driver posture (10:10 o'clock position) and let free during at the impact. For the tests at 30 kph the steering wheel was removed and the hands were maintained in a position corresponding to the natural posture for a driver during the acceleration phase of the sled and left free during the impact. For this purpose, the hands were fixed with two vertical nylon wires. The wires were released at the impact with a mechanical opening. The position of the arms (identified by the position of some particular anatomical points: acromion, lateral humeral epicondyle and radial styloïde) was closed to the position with the steering wheel. These positions were recorded with a 3D measurement device ("FARO arm").

Test Subjects – PMHS were provided by the department of Anatomy of Medical University of Lyon. The tested subjects were six recent unembalmed corpses of men who have given voluntarily before dying their body to Science. All tests were in accordance with the French ethical rules for the biomechanical experimentation (Got, 2001, Devers, 2001).

Subjects were examined for HIV and Hepatitis or other infectious diseases, anatomic abnormalities or signs of very long decubitus. They were chosen as closely as possible to the 50th percentile human body. The anthropometries of the tested subjects are given in the Table 2. In order to simulate living conditions, pulmonary pressurisation was performed prior to testing but the pressure was not recorded. No vascularisation was performed.

Table 2 – Post mortem human subject anthropometry characteristics

Test	Age	Weight (kg)	Height (cm)	Sitting height(cm)
FID11	46	63	183	100
FID12	83	69	168	92
FID13	74	67	168	89
FID14	78	82	180	93
FID15	81	58	167	87
FID16	90	45	177	94

Testing Device– A catapult was used to simulate impact, where the boundary conditions (initial speed, deceleration law and stopping distance) were well known. A cable winch coupled with an inertia flywheel propelled a rail-guided carriage. A PC Computer controlled the operation. The carriage was propelled against a concrete wall, and was stopped by a shock-absorbing system composed of polyurethane tubes.

The sled environment was set up according to the needs expressed by the FID consortium (Vezin, 2000. The seat geometry was close to those of a standard mid-size car and was fixed at the same position for each test and was independent of the anthropometry of the PMHS. The seat has a slope of 18° degrees and the footrest a slope of 43° degrees. The feet of the surrogate were fixed on the footrest.

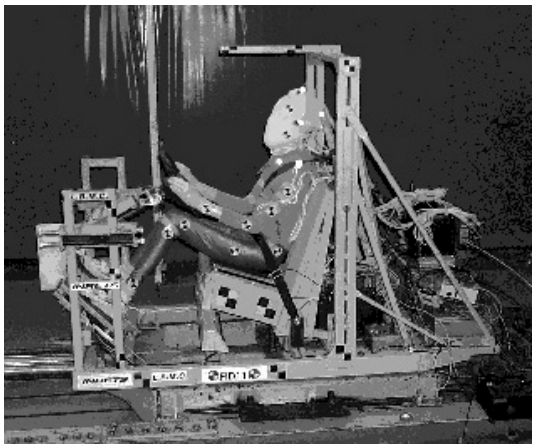

Fig. 2 – Rigid seat for the sled tests with post mortem human subject.

INSTRUMENTATION AND DATA PROCESSING

Subject instrumentation – Tri-axis accelerometers were mounted on the thoracic spine (T1, T8 and T12 vertebrae) and on the sacrum (Fig. 3.). A magnetohydrodynamic angular rate sensor was also mounted on the sacrum to record the angular velocity in the transverse direction. Concerning the Thorax, tri-axis accelerometers were mounted on the upper and lower part of the sternum; mono-axis accelerometers were also fixed on the 4th and 6th ribs on each side of the thorax. The left and right upper arms and acromions were equipped with tri-axis accelerometers (Fig. 3).

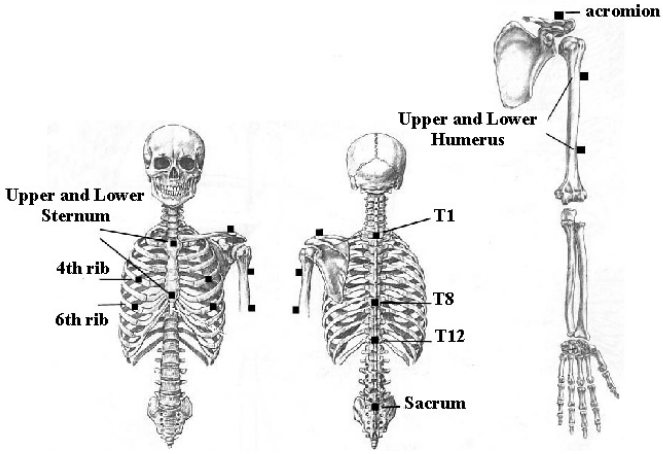

Fig. 3 – Position of the accelerometers on the skeleton

The sign convention and standard instrumentation used was the SAE J211 convention (March 1995) for all the tests. In the following, the x (or longitudinal) and z (or vertical) direction means the moving axis of the accelerometer during the crash. The z-axis is positive downwards and the x-axis is positive in the crash direction. Concerning the head, the Frankfurt plane defines the x and y-axis. The line between the centre of gravity and the midpoint of the line connecting the infraorbital notches defines the positive x-axis. The y-axis is positive towards the right ear.

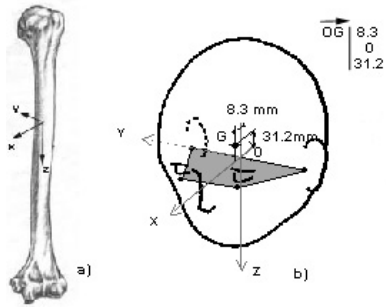

Fig. 4 –- Description of the anatomical reference,
a) Left humerus; b) Frankfurt plane and associated reference
(O origine of the reference, G centre of gravity of the head

Based on the N×1 method (Oudenard et al., 1991), INRETS-LBMC had developed a specific rig, weighing 330 gr, equipped with 12 accelerometers allowing the measurement of the six components of the acceleration of the head's centre of gravity (Bruyere, 2001). The resultant was calculated from the three components of the acceleration at the centre of gravity G (Fig. 4) defined by the mean position described by Beier et al. (1980).

Each component of acceleration data was acquired according to the SAE-J211 regulation and all data were acquired at 10,000 samples/sec. All the data were numerically re-filtered with a low pass band (60 Hz) digital Butterworth filter, prior to compute resultant accelerations.

Response procedures – Due to the variability in subject geometry and inertial properties, the subject responses were normalised to the standard anthropometry of the 50th percentile male. The normalisation procedures were described in details by Eppinger et al. (1984). The scaling factor λ based on the subject mass (M) in kg is shown in equation (1.). The accelerations, times and forces can be expressed in terms of initial parameters, denoted with subscript i, and the scaling factor (Eq. 2 to 4).

$$\lambda = (75/M_i)^{1/3} \quad (1)$$

Acceleration: $A = A_i/\lambda$ (2)

Time: $T = \lambda \times T_i$ (3)

Force: $F = \lambda^2 \times F_i$ (4)

Caution is necessary when applying this scaling method, because this standard scaling procedure is based on the assumption that all PMHS have the same mass density and the same modulus of elasticity. This hypothesis is open to criticism. The biomechanical properties depend, for example, on the age and the genre of the subject.

Due to anthropometrics and instrumentation mounting differences between PMHS and due to the unmeasured angular accelerations, only the linear acceleration resultant can be used for the comparison between the tests. Nevertheless the comparison of the components of the acceleration could give important information on the behaviour during impact and are sometimes discussed.

RESULTS

INJURY

Pre-test radiographies were conducted to identify any existing injuries or anomalies, but also to verify the mounting and the location of instrumentation. Following the tests, a pathologist and an autopsy specialist performed standard autopsies. Examinations of the cadaver's abdomen, viscera, head and neck, spine, and other skeletal elements were performed. The number and the locations of rib fractures were documented. All injuries were coded according to the Abbreviated Injury Scale (Faverjon et al., 1994) and the maximum AIS value (MAIS) was determined. AIS was assigned to rib fractures as follows: 1 rib, AIS 1; 2-3 ribs AIS 2; >3 ribs on one side and ≤3 ribs on the other side, AIS 3; >3 ribs on each sides with stable chest wall, AIS 4.

The thoracic injuries concern ribs, clavicle and sternum. Table 3 summarises the numbers and location of fractures. For one test at 50 kph, the left side of the thorax is massively injured. It roughly coincides with a line from the sternum to the left of the chest. The number of fractured ribs is considerably reduced for the two other tests at 50 kph. No head or neck injuries are identified. No other injuries, in particular visceral injuries are found for all cadavers.

For all test conditions, the left ribs are more often fractured (Table 3). Sternal fractures are also observed. Concerning the clavicle, no fractures are observed but only dislocation of the sternoclavicular joint occurred. These dislocations appear on the left shoulder and they are probably caused by the shoulder belt.

The injuries are a little reduced with the reduction of the velocity. No injuries are found for the clavicle for all the tests at 30 kph and the right side of the thorax have encountered any fractures, except one rib for one test.

Table 3 – Injury Summary Table, (Rib number $^{\text{number of fractures for the rib}}$)

Test	Left ribs fractured	Right ribs fractured	Sternal fracture	Clavicle fracture
FID11	2^2, 3^2, 4^2, 5, 6, 7, 8, 9	no	no	sternoclavicular dislocation (Left)
FID12	3, 4,	4, 5^2, 6	yes	no
FID13	no	no	no	sternoclavicular dislocation (Left)
FID14	3, 5	no	yes	no
FID15	2^2, 5	6	no	no
FID16	no	no	no	no

The Table 4 gives the AIS and MAIS code for the injured parts. The AIS for the thorax is lightly lower for the tests conducted at 30 kph than the 50 kph tests. But if we consider the severity of the impact at 50kph, the effect of the combination of the airbag and the shoulder belt is comparable as a decrease of the severity (velocity).

Table 4 – Abbreviated Injury Scale Summary Table

Test	AIS Thorax	AIS Sternum	AIS Clavicle	MAIS
FID11	3	0	1	3
FID12	2	1	0	2
FID13	0	0	1	1
FID14	2	1	0	2
FID15	2	0	0	2
FID16	0	0	0	0

BELT LOADING

The loads in the shoulder and lap belts are presented respectively in the Fig. 5 and Fig. 6. The activation of the load limiter occurs only for the tests at 50 kph and occurred rapidly (50 ms) due to the movement of the body in the front direction. The maximum of the pelvic belt is in phase with the shoulder belt. More differences are observed between the tests at 30kph, the combination of the airbag and the force-limited belt has a greater influence on the movement of the body than a single belt. The peak of the pelvic belt appears later, 70 ms instead 50 ms. The kinematics of the body is better controlled by this restraint system and the influence of the other parameters is lessened.

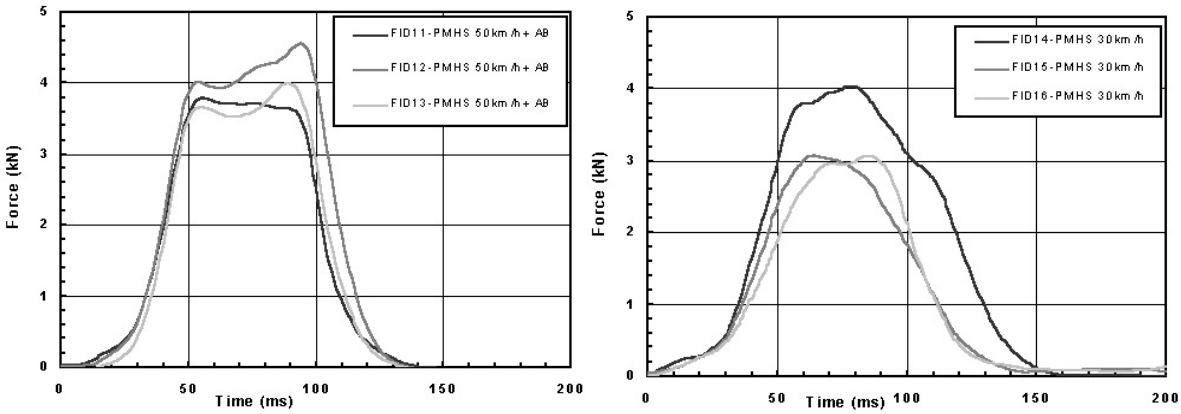

Fig. 5 – Shoulder belt force time history Left) 50 kph, FL Belt + Airbag; Right) 30kph, FL Belt

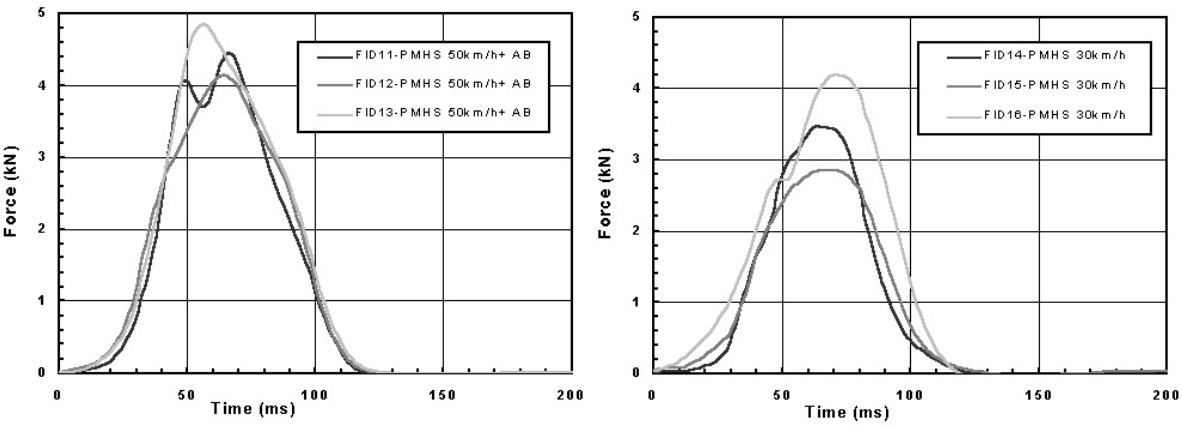

Fig. 6 – Lap belt force time history Left) 50 kph, FL Belt + Airbag; Right) 30kph, FL Belt

SUBJECT ACCELERATION

Figures 7 to 17 present and compare the response of the resultant acceleration for the two test conditions. There is a good test-to-test repeatability despite the differences not accounted for by the scaling procedure.

Thoracic and lumbar spine – The subject resultant acceleration curves exhibit waveforms with one major maximum. A first lower peak is also visible on the chest (T8 and T12) resultant for the tests with airbag (Fig. 8 and 9). The first maximum occurs roughly at the time of the peak shoulder belt (50 ms) and the second one coincides with the instant as the subject loaded the airbag (80 ms). The T1 acceleration does not captured this bimodal response (Fig. 7). The single peak for the T1 resultant occurs also at the time of the impact with the airbag. In return, the maximum resultant for

the sacrum is in phase with the maximum of the pelvic belt load. The single maximum on the curves of the tests without airbag coincides with the maximum of the belt load.

The duration of the acceleration is different between the two series of tests. If we look at the interval time where the acceleration was greater than $0.25 \times A_{max}$, at 50 kph, the duration of the deceleration is about 80 ms for all the curves when at 30 kph we can estimate the duration around 110 ms. Except for the sacrum resultant (Fig. 10), where the pulse duration is quite similar, it is probably due to the effect of the separated pelvic belt.

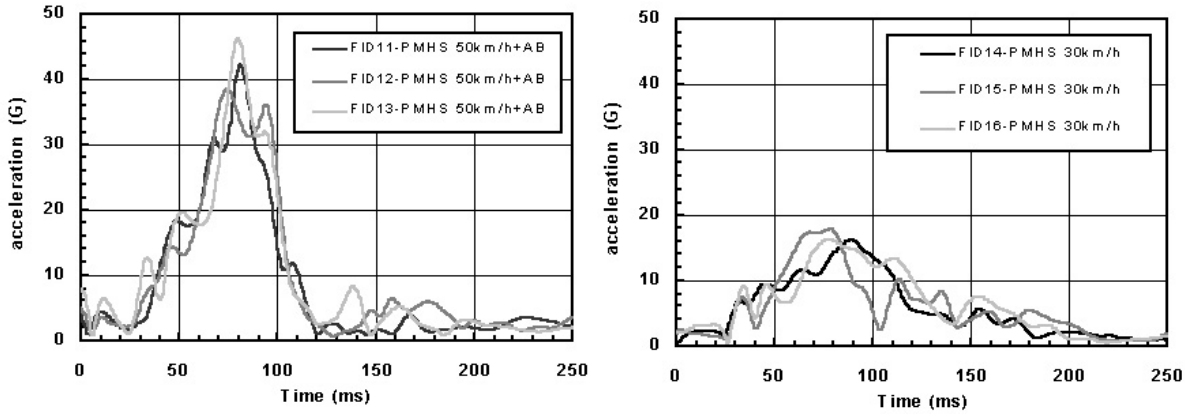

Fig. 7 First thoracic vertebrae (T1) resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

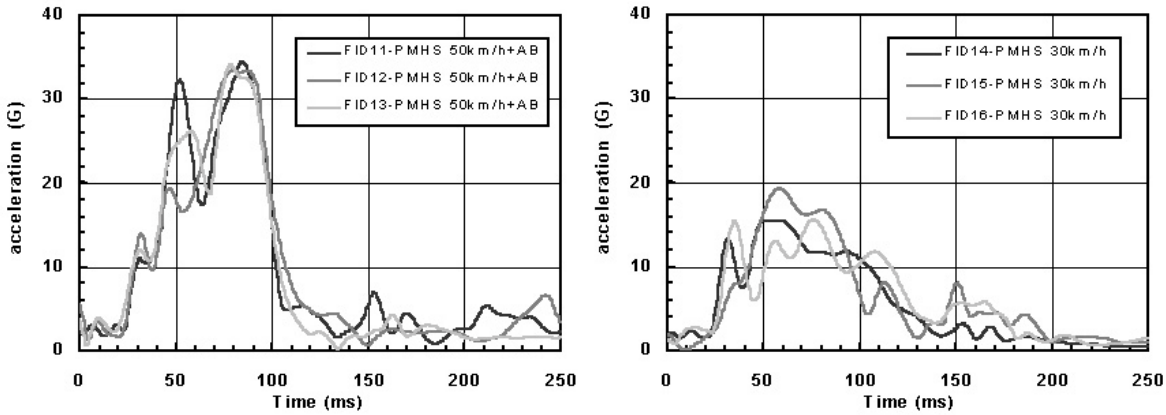

Fig. 8 Eight thoracic vertebrae (T8) resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

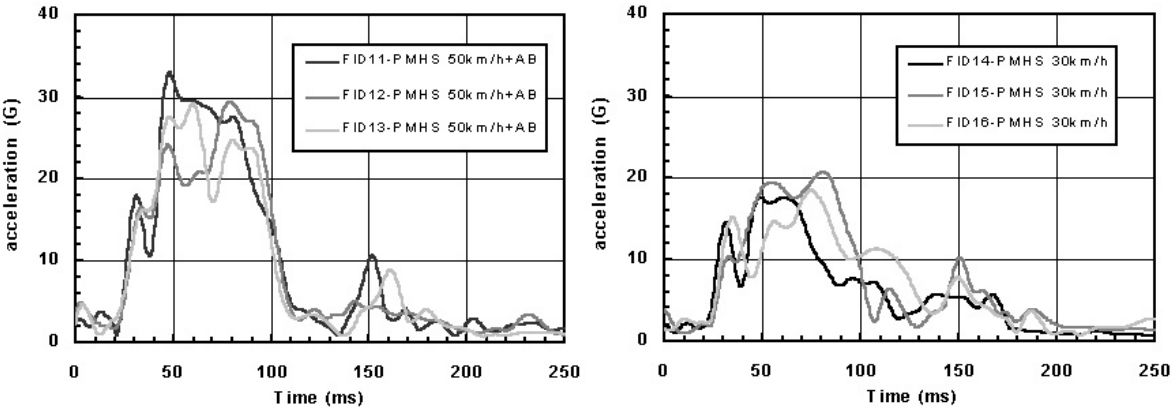

Fig. 9 Twelfth thoracic vertebrae (T12) resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

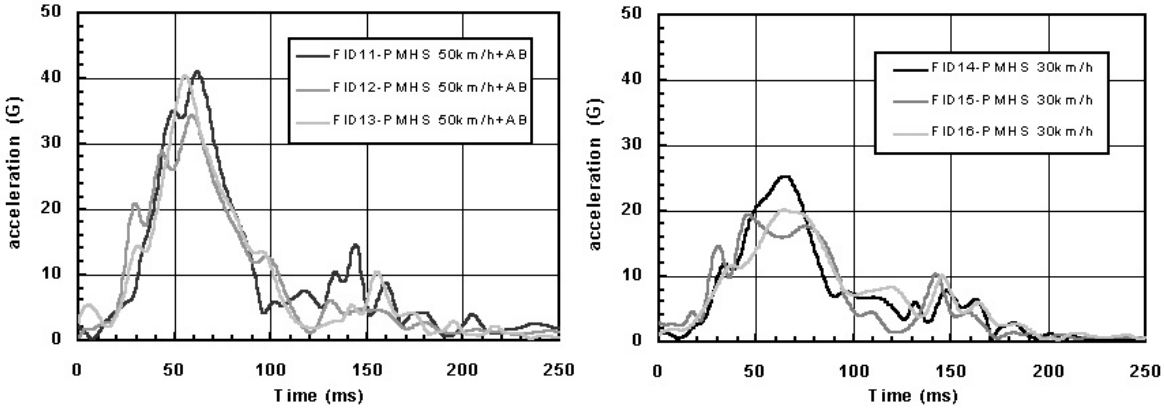

Fig. 10 Sacrum resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

The Figure 11 shows clearly that the behaviour depends on the speed and restraint systems. Without airbag, the belt with a 4kN load limiter less restrains the body. The lower deceleration is quasi-uniformly distributed along the thoracic spine and the movement of the corpse is quasi-uniformly in the front direction of the impact since the vertical components are generally small compared the horizontal one. Whereas, with airbag the value of the maximum deceleration along the spine is very different. Specifically, the upper thorax (T1) is more decelerated than the lower thorax (T12). As a result, interactions with the lower steering wheel rim and airbag can occurred. The contribution of the vertical acceleration is in this case of loading not negligible and particularly around 80-100 ms. This time corresponded to the instant when the chest has bumped into the airbag. The deceleration of the pelvis is higher and approximately at the same level of the T1 deceleration, but the influence of a separate pelvic belt compares to a standard 3-point belt is not studied.

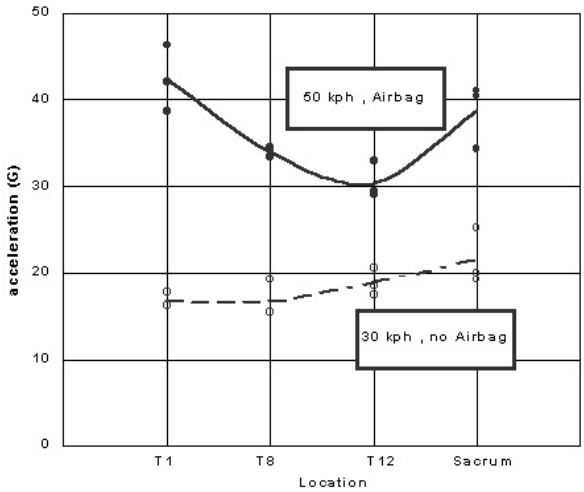

Fig. 11 – Comparison of the maximum resultant acceleration at different location along the spine

Sternum – The two peaks waveforms are also observed for the sternum resultant during the test with airbag (Fig. 13 and 14). The instant of the 1st peak and, at a less point, the 2nd peak appears earlier compare to the spine acceleration curves.

For the lower sternum; which coincides with the T12 vertebra; the maximal deceleration is higher for the sternum than for the thorax. It is a consequence of the chest deformation caused by the restrained system. The differences are not so obvious for the tests without airbag. In consequence, the thoracic deformations are probably lower even if the thoracic injuries are quite similar or little reduced. This observation confirms that the single thoracic deflection is not the only predicting parameter for the thorax injuries. Concerning the duration of the acceleration pulse the same behaviour as for the thoracic spine is observed.

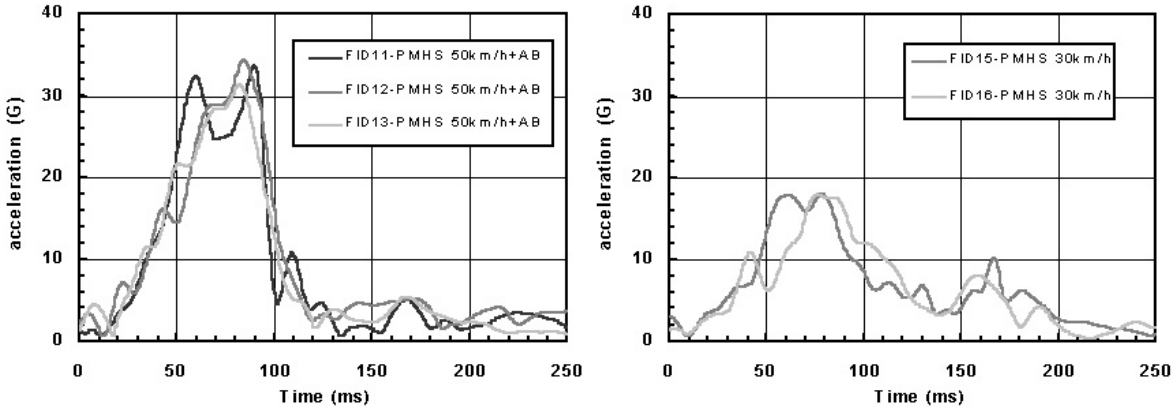

Fig. 12 sternum (upper part) resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

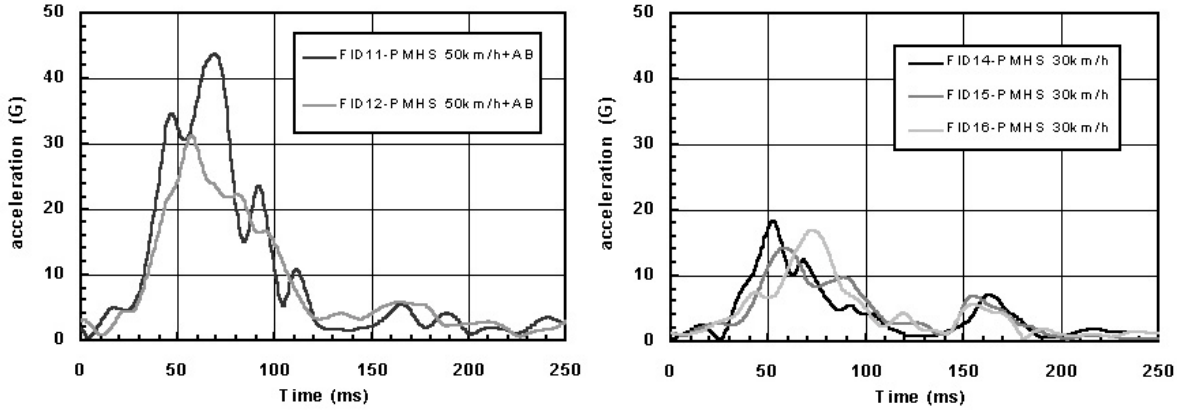

Fig. 13 sternum (lower part) resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

If we compare the lower and the upper acceleration, the two curves for the tests with airbag are similar. On the other hand, the maximum on the curves for the tests without airbag do not appeared at the same time. Approximately 50 ms for the lower part and 70-80 ms for the upper part of the sternum. The load of the chest by the airbag is more uniform than the load by a single belt.

Shoulder -. The behaviour of the shoulder was studied through the acceleration of the acromion and the humerus (upper and lower part).

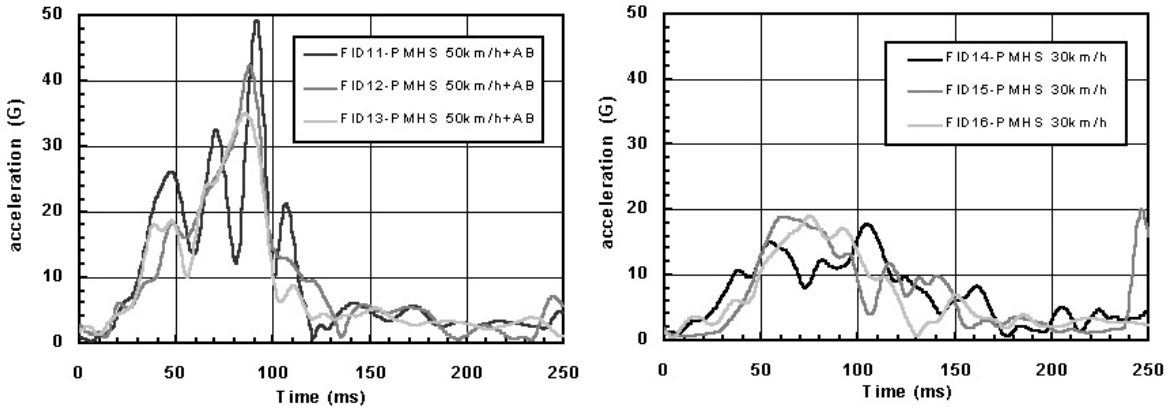

Fig. 14 Left acromion resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

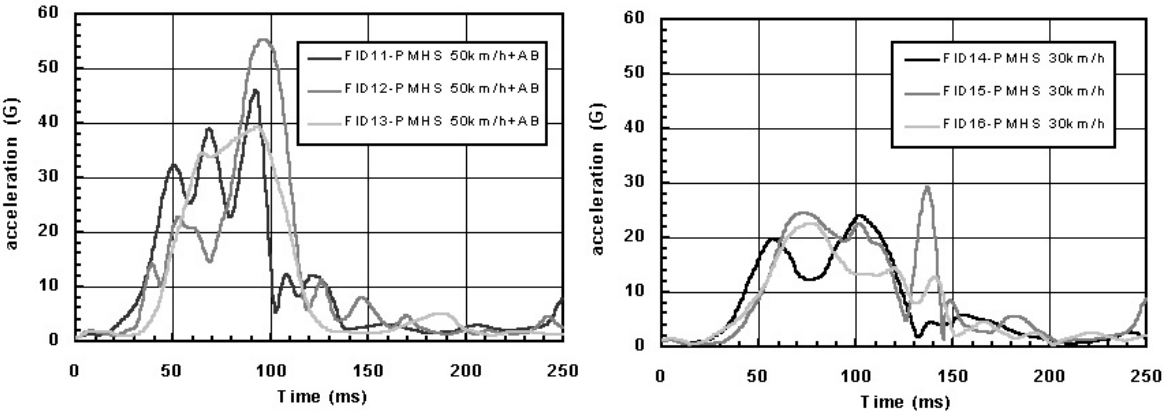

Fig. 15 Left humerus resultant acceleration (upper part) time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

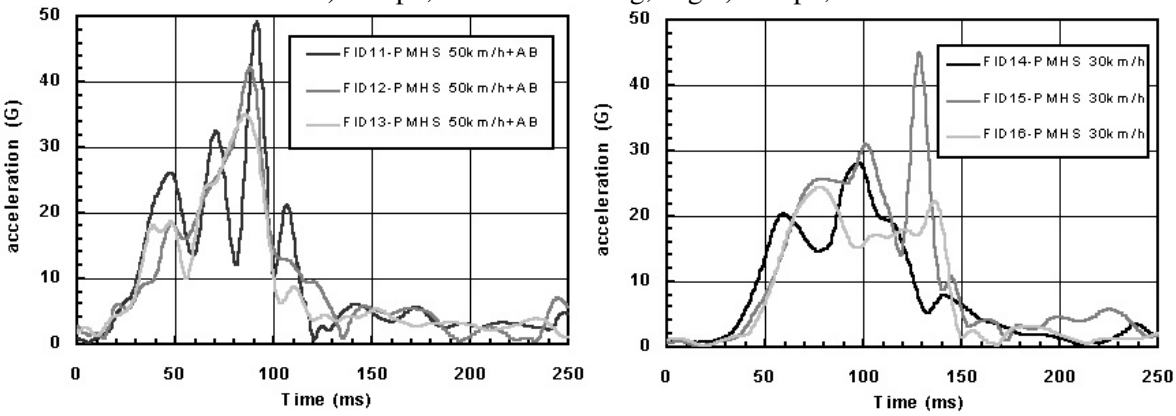

Fig. 16 Left humerus resultant acceleration (lower part) time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

The bimodal behaviour, already showed for the torso behaviour, is visible for the shoulder during the airbag tests. The resultants increase rapidly up to the activation of the load limiter. After that, the slope changes and the increasing of the resultant is lower up to the load by the airbag. It is the same for the humerus. For the test without airbag, the first part of the curves has the same behaviour but the resultant remains constant during the impact. It could be interesting to evaluate the contribution of each component, linear and angular, of the humerus resultant. A tentative of calculation of the angular acceleration with the 6 accelerometers fixed on the humerus is in preparation and is not presented in this paper.

Head – The head resultant acceleration at the centre of gravity of the head is calculated with the measurement of 12 accelerometers fixed on the rig screwed on the skull. The two series of curves have the same single peak waveform. The maximum acceleration corresponds respectively to the load by the airbag (50 kph) and to the maximum load by the shoulder belt (30 kph). A greater dispersion is observed for the test without airbag, particularly for the value of the maximum. It is certainly due to the differences more important differences of anthropometry for this series not take into account by the scaling procedure.

The head injury criterion HIC at 35 ms is determined by the following equation (Eq. 5).

$$HIC = \left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} \gamma(t)\,dt \right)^{2.5} (t_2 - t_1) \qquad (5.)$$

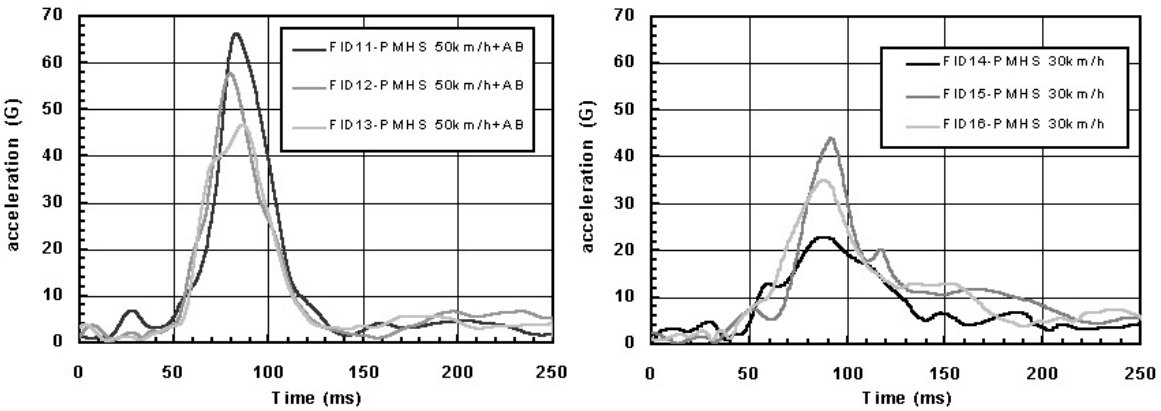

Fig. 17 Head resultant acceleration time history
Left) 50 kph, FL Belt + Airbag; Right) 30 kph, FL Belt

The maximum accelerations and HIC values are collected in Table 6. In all cases the HIC is lower than the current level of HIC = 1000, used in the risk curves but it is difficult to conclude in terms of risk injury because there was no steering and consequently no impact for the lower velocity tests.

Table 6 – Head Injury Criterion and maximal resultant acceleration

Test	HIC	Acceleration (G)
FID11	620	66.4
FID12	403	57.7
FID13	350	46.7
FID14	66	22.9
FID15	210	44.0
FID16	160	34.9

DISCUSSION

This study aims to provide new data concerning human thorax/shoulder complex behaviour during sled tests with different case of shoulder loading. In order to use these data for the evaluation of the Hybrid III and THOR-alpha dummy, the test configuration was simplified, compared to a real car crash situation. The seat was rigid and its position and orientation were the same for each test independently of the anthropometry of the subject. Moreover, the lap and shoulder belts were separated. The effect of different restrain system: Airbag with load limited shoulder or only load limit shoulder belt had been investigated.

Caution in applying results obtains with PMHS to living persons is necessary, namely due to the lack of muscle tone in PMHS. However, *"the lack of muscle tone is not a serious drawback because muscular response usually occurs too late to affect body kinematics in a crash"* as King and Viano wrote (1995). Moreover, variability in the PMHS behaviour cannot be avoided due to many parameters such as anthropometry and properties of the biological tissues. In order to limit this variability, subjects were selected as closely as possible to a 50th percentile anthropometry and the PMHS results were normalised.

The standard scaling procedure based on the mass of the surrogate has already been discussed in a previous paragraph, but it will be necessary to specify some points about this procedure. The two main assumptions to obtain the scaling parameter are that the surrogates have the same density and the same modulus of elasticity. It is well known that these two parameters are dependent on the age, gender, morphology, pathology and/or many others parameters. It has also been often stated in many studies (Kallieris et al., 1995; Shaw et al. 2001 etc.), that, in some case, the mass-scaled data were more dispersed than the unscaled data. Our opinion is that, when the weights of the surrogates are very different (for instance, our tests at 30 kph), the scaling procedure, even if it do not take into account all the subject parameters, provide an improvement for the comparison of the tests. On the other hand, when the surrogates are close to each other, (for example our 50 kph tests) the errors introduce by the hypothesis stated above are more important than the eventual benefit.

Considering these difficulties in studying human behaviour, the results presented in this paper will allow us to evaluate the influence of the restrained system of a car occupant on his behaviour during a crash. These results will also allow us to compare the Hybrid III or THOR-alpha dummy with the PMHS behaviours in a particular loading case and to assess the biofidelity of the dummies (Vezin et al., 2002).

The number of fractures is sometimes important despite the reduction of the load on the thorax due the belt load limiting system and the Airbag. It is not surprising if we keep in mind the great severity of the impact and this observation is correlated with the results obtain on the same test device without airbag during the HUMOS experiments (Vezin et al., 2000, 2001). A recent study (Ydenius and Kullgren, 2001) has shown that the injury risk for MAIS>2 is greater than 80% for a mean sled deceleration close to our tests conditions (>15 Gs). This observation can be correlated with a injury risk around 60% at 50 kph (MAIS>2) and 20% at 30 kph (but with a lower deceleration). Although, the bone injuries observed on PMHS were often greater than those observed in real car crashes, these results are in accordance with the data presented here. Nevertheless, the reduction of injuries thanks to the airbag is not negligible. The reduction of thoracic injuries is comparable to the reduction obtain with lessening the severity of the impact (i.e.: lower velocity, lower energy).

The occupant restrain systems are enforced by the requirements to reduce thoracic injuries. In order to evaluate the capacity of the restraint systems, thoracic injury criterion have been developed. These criterion have been based on the chest deflection, on the rate of chest compression (V*C) or on the acceleration of the centre of gravity of the chest. A more precise criterion developed in the specific case of sled tests have been proposed by NHTSA's researchers (Kuppa and Eppinger, 1998). This thoracic criterion, CTI, is based on a two parameters logistic regression:

$$CTI = \frac{A_{max}}{A_{int}} + \frac{C_{max}}{C_{int}} \qquad (6)$$

where A_{int} and C_{int} are respectively chest acceleration and maximal deflection normalization constants defined for various dummy sizes.

However, such criterion must differentiate the behaviour of the thorax for the set of loading conditions that cover the extent of restraint systems. Previous studies have demonstrated that the current thoracic injury criterion (Kent et al., 2001; Shaw et al., 2001) do not represented with sufficient efficiency and sensitivity the influences of the type of restraint systems used. The acceleration measured at the chest for the dummy has been used as injury predictor, but this injury criterion has been based on the T1 measurement on cadavers. The T1 acceleration is considered to be representative of the chest acceleration. The present study confirms the observation made for other cadaver tests, which included accelerometers at both T1 and T8 vertebra, that this assumption is not true. For instance, for some conditions, the T1 acceleration does not necessarily represent the chest acceleration. The maximum value of the T1 acceleration measured in these different configuration tests does not, indicate the relative loading of the belt and airbag because, the peak acceleration occurred only when the subject was load by the airbag. The chest (T8) acceleration, by exhibiting a bimodal behaviour, seems to give more information about the thorax loading conditions for driver sled tests and would be more predictive parameters. Kent et al. (2001) have yet demonstrated that it is not he case for right front passenger sled tests. These results are not in opposition with the present data. The fact that different studies with different kinematic conditions and different restraint systems show different and contrary observations, demonstrate the importance of the crash conditions.

A predictive injury criterion has to be sensitive to the impacts boundary conditions. The bimodal waveform of the chest deceleration suggests that the injury may be occurred at the same time, or be a consequence, of one or both peak of acceleration. The correlation of the injury, probably related to the chest compression, with the corresponding peak of acceleration is one of the possibilities to improve the injury criterion. Other possibilities are the acceleration measured in two point, T1 and T8, or T8 and sternum; correlated with the thoracic deflection or the rate of thoracic deflection will be more precise. The contribution of the component, better than the linear resultant, could represent the dynamics of the thorax that lead to an injury. In particular, the torso rotation, even if the restraint systems limit her influence, should be taken into account.

The tests on cadavers provide principally information on the mechanism of the hard tissues (bones) but not for the soft tissue. An alternative approach could be the numerical modelisation in order to estimate the injury mechanisms. In opposition of the existing criterion based on statistics, this approach is based on mechanical criterion related to the stress inside the tissue and which permit to explain the injuries.

The recent experimentations in biomechanic have especially allow to better understand the mechanical behaviour of the whole body and/or some body parts. Many European projects have provided characteristics curves (Force-displacement, acceleration time history, for example) for the development of new dummies. Within the FID project, reviews have also been provided in order to obtain results that can be really. In other words, results where the tests conditions are well known. Biomechanical research concerning the behaviour of the shoulder during frontal impact conditions is rare, and the data presented here would contributed to reduce the lack of biomechanical data which can serve as the basis from which biofidelity requirements for the shoulder can be defined.

All the data presented in this paper would serve to define response corridors for the evaluation of the THOR-alpha and Hybrid III dummies. The envelope of the experimental tests will define the corridor, in accordance with the FID consortium. The biofidelity of the dummy will be assess with regard to these corridors, and improvement will be proposed to get a dummy with more humanlike behaviours for the different tests conditions presented in the present study.

CONCLUSIONS

Two series, each comprising three cadavers tests, were carried out at INRETS - LBMC. The first series performed at 50kph used a seat equipped with a 4 kN load-limited shoulder belt coupled with an airbag and the second series carried out at 30 kph used a single 4 kN load-limited belt.

As expected, results show that the type of restraint systems and the test conditions modify the behaviour of the PMHS. The restrained conditions are not reflected in the same way for different segment of the body. For instance, the influence of the airbag deployment is well observed for the chest (T8 and T12) spine behaviour, but is not recorded by the instrumentation mounted on the first thoracic vertebra. Injury criteria or risk curves that would predict the injury than can occur in car crash have to take into account this observation. The new criteria named CTI proposed by the NHTSA and based on a linear combination of the T1 maximum resultant acceleration and the maximum chest deflection is a step in this direction. But this study suggests that the chest acceleration (T8) could be a better parameter than the T1 resultant. Probably, the study of the influences of the components of the acceleration rather than the resultant should be necessary. A parameter, which could take into account these influences, should be a more predictive parameter.

It is not easy to obtain experimentally the whole description of the acceleration (the linear and angular components) and simultaneously the chest deformation. The authors do not know the existence of any completed studies about this subject. For that numerical models, which would reproduce the human behaviour whatever the load levels and the impact conditions, have to be developed and improved. The basic assumption of the European project Human Models for Safety 2, **HUMOS 2**, is that a biofidelic model shall be structurally very close to the real human body. This assumption means that a correct representation of the main human structures is needed, the skeleton, but also the main organs and muscles. Scaling (to simulate different size of occupant) tools and positioning (to simulate different restrained conditions and/or crash car situation) tools have to be developed. A biomechanical knowledge, particularly on the effect of the muscle tone and pressurisation, is also needed in order to validate the model with a realistic behaviour in a car crash situation. The data obtained from the cadaver tests presented in this paper and completed with more tests on the same sled, will be used as reference data for the validation of this numerical model of the human body and will be included in the future European Biomechanical Database.

Finally, the results presented here are also a part of the data collected for the evaluation of the Hybrid III and the new THOR-alpha during the **FID** project. The response corridors of behaviour obtained from this data are used to evaluate the biofidelity of both dummies (Vezin et al., 2002) in order to propose improvement of the THOR-alpha.

ACKNOWLEDGEMENTS

The FID project is funded by European Community (DG TREN) under the 'Competitive and Sustainable Growth' Programme and co-ordinated by TNO Automotive (NL). The partners involved in the FID program are : Transport Research Laboratory (TRL, UK), Polytechnical University of Madrid (S), University of Heidelberg (D), Bundesanstalt für Strassenwesen (BASt, D) and Institut National de Recherche sur les Transports et leur Sécurité (INRETS, F).

The authors would like to thank to the Doctor Eric Voiglïo (surgeon of the Hospital of Lyon) from the University of Medicine, Claude Bernard Lyon, and the Doctor Michelle Ramet from the INRETS-LBMC for their contributions to the PMHS experiments and the autopsy and the injury description.

The authors would like to thank Pierre Lapelerie, Michel Marais, Sophie Serindat, Gérard Goutelle, Marcel Callejon and Alain Gilibert, for doing such a good job in collecting data, photos, measurements, during the experiments.

REFERENCES

ADRIA Partners, "Advanced crash Dummy Research for Injury Assessment in frontal test conditions", PL96-1074, Rapport final, 10 January, 2000.

Beier G., Schuller E., Schuck M, Ewing C.L., Becker E.D., Thomas D.J., "Center of gravity and moments of inertia of human heads", Vth International IRCOBI conference, 1980, pp 218-228.

Bruyere K., Bermond F., Bouquet R., Caire Y., Ramet M., Voiglio E., "Human maxilla bone response to 30° oriented impacts and comparison with Frontal Bone Impacts ", 44th Annual Scientific Conference of the AAAM, Chicago, Illinois, 1-4 October 2001, pp 219-234.

Devers G., "Juridical statute of the human deceased corpse in France" Workshop on Ethical Guidelines in Biomechanical Testing, European Passive Safety Network, INRETS, Bron, France May 23-23 2001, pp 94-100.

vanDon B., vanRatingen M., Bermond F., Masson C., Vezin P., Hynd D., Owen C., Martinez L., Knack S, "Biofidelity impact response requirements for an advanced mid-sized male crash test dummy" EEVC/WG12, IRCOBI 2002 (accepted)

Eppinger R.H., Marcus J.H., Morgan R.M., "Development of Dummy and Injury Index for NHTSA's Thoracic Side Impact Protection Research Program", SAE Publication N° 840885, Government /Industry Meeting and Exposition, Washington, D.C, May 21-24 1984.

Faverjon G., Foret-Bruno J.Y., Got C., Morvan Y., Lepage M., Portier L., Thomas C., Vallee H., "Abbreviated Injury Scale - 1990 Revision, abbreviated impairment scale - 1994 Revision French translation", June 1994.

Got C., "Ethical issues in the use of cadavers for impact tests" Workshop on Ethical Guidelines in Biomechanical Testing, European Passive Safety Network, INRETS, Bron, France May 23-23 2001, pp 31-35.

Horsch J.D. Schneider D., "Biofidelity of the Hybrid III Thorax in high velocity frontal impact", SAE No. 880718, Society of Automotive Engineers International Congress and Exposition, February, 1988.

Kallieris D., "Progress Report WP3 UH-31-001", FID project technical document, 2000.

Kallieris D., Rizzetti A., Mattern R., Morgan R., Eppinger R., Keenan L., "On the synergism of the driver air bag and the 3-point belt in frontal crashes", Proceedings of the 39th STAPP Car Crash Conference, N° 952700, San Diego, California, November 8-10 1995, pp 389-401.

Kallieris D., University of Heidelberg, Germany, Personal communication, 2001.

Kent R., Crandall J., Bolton J., Duma S., "Driver and right-front passenger restraint system interaction, injury potential, and thoracic injury prediction", 44th Annual Scientific Conference of the AAAM, Chicago, Illinois, 1-4 October 2001, pp 261-282.

Kent R., Bolton J., Crandall J., Prasad P., Nusholtz G., Mertz H., Kallieris D., "Restrained Hybrid III dummy-based criteria for thoracic hard-tissue injury prediction", IRCOBI Conference, Isle of Man, United Kingdom, October 2001, pp 215-232.

King A., Viano D.C., Mizeres N., States J.D., "Humanitarian benefits of cadaver research on injury prevention", The journal of Trauma: Injury, Infection and Critical Care, Vol. 38, No.4, 1995, pp 564-569,.

Kuppa S.M., Eppinger R.H., "Development of an improved thoracic injury criterion", Proceedings of the 42nd STAPP Car Crash Conference, N° 983153, Tempe, Arizona, November 2-4 1998, pp 139-153.

Shaw G., Crandall J., Butcher J., "Biofidelity of the THOR advanced frontal crash test dummy", IRCOBI Conference, Montpellier, France, September 2000,.pp 11-29.

Shaw G., Kent R., Sieveka E., Crandall J., "Biofidelity of the THOR advanced frontal crash test dummy", IRCOBI Conference, Isle of Man, United Kingdom, October 2001, pp 241-254.

Oudenard L., Bendjellal F., Bellini A., Uriot J., "The use of a multi-accelerometric method in automotive safety tests", 13th ESV Conference, Paris, France, November 1991, pp 862-871.

Vezin P, Bruyere K., Verriest J.P. "Comparison of the biofidelity of the Hybrid III and THOR-alpha response to frontal sled test", 46th STAPP Car Crash Conference, 2002 (submitted).

Vezin P., Bruyere-Garnier K., Bermond F., "Comparison of head and thorax cadaver and Hybrid III responses to a frontal sled deceleration for the validation of a car occupant mathematical model", 17th ESV Conference, Amsterdam, The Netherlands, June 2001, paper number ID#: 114.

Ydenius A., Kullgren A., "Injury risk functions in frontal impacts using recorded crash pulses", IRCOBI Conference, Isle of Man, United Kingdom, October 2001, pp 27-38.