HUMAN SPINE BEHAVIOUR UNDER THORACIC AND PELVIC LATERAL IMPACTS COMPARISON WITH WORLDSID DUMMY BEHAVIOUR

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ABSTRACT

Human spine behaviour under lateral impacts was studied through Post Mortem Human Subject (PMHS) experiments carried out within the European project SIBER (SIB, 2000) and previous tests conducted in a French PREDIT Program (BOUQUET, 1994). The aim of these experiments is to characterise the coupling between the upper and lower torso created by the lumbar spine. For this purpose, PMHSs were struck to the thorax and then to the pelvis at different velocities: 5 thoracic impacts were conducted at 4 m/s, 5 pelvic tests at 6.6 m/s. In addition, previous tests carried out in similar conditions were re-analysed (5 thoracic impacts at 3.3 m/s, 5 others at 5.5 m/s and 10 pelvic impacts at 6.6 m/s). Impactor force, subject accelerations at T1, T4, T8, T12 and sacrum were recorded, while 2D motion of the subject's spine was filmed. Two WorldSID dummies (prototype and pre-production version) were submitted to the same type of impacts to assess their lumbar spine behaviour.

The dummy responses were evaluated against corridors defined from PMHS responses. The WorldSID dummy showed an improvement with respect to the current European regulatory dummy. Its lumbar spine allowed a more biofidelic coupling between its upper and lower torso which created a more human-like kinematics.

Key Words: Biomechanics, Cadavers, Crash test dummy, Lumbar spine, Side impact.

THE LUMBAR SPINE IS RESPONSIBLE FOR THE MECHANICAL COUPLING between the thorax and the pelvis. It is also one of the most flexible parts in the vertebral column. Thus it plays an important role in the occupant kinematics during a crash. However, very few studies are available on this topic and no biofidelity requirement is given for dummy lumbar spine behaviour in side impacts. Depending on the lumbar spine characteristics, the thorax and the pelvis can be loaded in a different way. With a stiff lumbar spine, any pelvic movement will produce a thoracic motion, whereas with a soft one, it will allow much more distortion in the spine before moving the upper torso. In general, the current dummy lumbar spine is made of a rubber cylinder crossed in its central part by a steel cable used to adjust lumbar spine stiffness during certification tests. This design allows the dummy to remain seated upright but doesn't necessarily give a humanlike behaviour in lateral shearing.

Being aware of the current dummy lumbar spine design limitations, the WorldSID lumbar spine has a completely new design for a more human-like response. It is made of a flexible rubber part in a U-shape, reinforced vertically by two rubber bars and attached by steel plates to the pelvis and the thorax (WorldSID user manual, 2003) (Fig. 23). A new design was necessary to improve kinematics of current dummies which behave almost as a single piece whereas PMHSs show less coupling in this region. However, no objective guideline on dummy kinematics exists (ISO, 1997). Within SIBER (SIBER, 2000), biofidelity targets have been proposed to evaluate dummy lumbar spines. For this purpose, localised impacts were delivered to the pelvis and the thorax of PMHSs and the coupling between the upper and lower torso was studied in these two different cases of impact.

METHOD

SUBJECTS – Post Mortem Human Subjects (PMHS), provided by the department of Anatomy of Medical University of Lyon, were fresh corpses of men and women who donated their body to Science. Traditional anthropometry measurements were carried out. Main subject characteristics are listed in Table 1.

Subject	Age	Sex	Weight [kg]	Height [cm]	Bitrochanter width [cm]	Sub-sternal thoracic width [cm]	Axiliary thoracic width [cm]
SIBER 01	77	F	67	161	34.5	28	29
SIBER 02	88	M	33	163	30.5	25.3	25
SIBER 03	79	F	52	159	31	27	26
SIBER 04	82	F	50	155	31.5	23	28.5
SIBER 06	94	F	50	148	31.5	25.5	29

Table 1 : Main subject characteristics

TEST SET-UP – The tests were conducted using a 23.4 kg rigid guided impactor with a circular impacting plate (∅ 150 mm). Different non-injurious and injurious velocities were chosen for the thorax and the pelvis. For the pelvic impacts, the centre of the impactor face was aligned with the greater trochanter. For the thoracic impacts, PMHS height was adjusted to align impactor face centre with the most lateral point of the thorax contour at the T8 level.

Subject instrumentation: Subjects were instrumented with tri-axial accelerometers (tri-axis) screwed onto several vertebrae: T1/T2, T4/T5, T8/T9, T12/S1 and sacrum. Data acquisition was made according to SAE J211 convention. All data were sampled at 10 kHz and filtered at CFC 180 Hz.

Subject photographic instrumentation: Photographic marker couples were mounted onto the accelerometer fixation plates or directly screwed into the bones for the lumbar vertebrae. Two cameras were placed behind the subject, one of them focussing on the lumbar spine region. The subject's instrumentation is shown in Fig. 1.

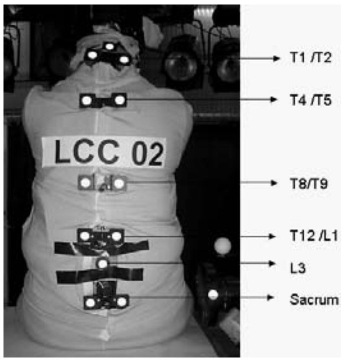


Fig. 1 : Subject's instrumentation and photographic markers

IMPACT CONDITIONS

Subject positioning: The subject was placed sideward in front of the impactor, upright seated on a table covered with two Teflon™ sheets and held in position by a cable attached to an electromagnet, released just before the impact. The time of release was adjusted depending on impact speeds. In general (except for SIBER 01), a pulmonary pressurisation was performed but without recording the pressure level. The lumbar spine tests followed four shoulder tests conducted at non-injurious level on the right PMHS side and at injurious level at its left. Thus, lumbar spine tests were conducted on the right side of the subject.

Test matrix: The SIBER test matrix is presented in Table 2. Tests on SIBER subjects (called LCC tests) were carried out at 4 m/s for thoracic impacts and 6.6 m/s for pelvic impacts. In addition, tests conducted within a previous national program were re-analysed for comparison. In this program, tests on 10 PMHSs (MRT subjects) were performed with a 23.4 kg impactor fitted with a 100×200 mm rectangular impacting plate. Some of these tests were thoracic impacts (called MRL tests) carried out at 3.3 and 5.5 m/s and some were pelvic impacts (called MRB tests) conducted at 3.4 and 6.6 m/s. In both cases, impact force, T1, T12 and sacrum accelerations were recorded. T1 measurements are not presented here and focus is made on the recordings obtained at T12 and sacrum levels which represent the upper and lower limits of the lumbar spine.

	Thoracic impact tests			Pelvic impact tests		
	Test	Subject	Velocity [m/s]	Test	Subject	Velocity [m/s]
SIBER Test Program	LCC 01	SIBER 01	4.1	LCC 02	SIBER 01	6.6
	LCC 03	SIBER 02	4.2	LCC 04	SIBER 02	6.6
	LCC 05	SIBER 03	4.2	LCC 06	SIBER 03	6.8
	LCC 07	SIBER 04	4.2	LCC 08	SIBER 04	6.9
	LCC 10	SIBER 06	4.2	LCC 11	SIBER 06	6.7

Table 2 : SIBER test matrix (LCC tests)

Post-tests: X-ray pictures of the pelvis and the thorax were taken to verify the location and the number of bone fractures. At the end of the tests, an autopsy was performed to carefully document rib and pelvis fractures. It should be noted that rib fractures could also be due to injurious left shoulder impacts.

DATA ANALYSIS – All presented data were filtered using a low pass band digital Butterworth filter (CFC 180). Impact force was given by the load cell placed between the impactor tube and its impacting plate. In the case of the SIBER tests, the force measurement was corrected by a factor equal to 1.088 because of the mass placed in front of the load cell. Subjects' responses were normalised according to Mertz's method (Mertz, 1984) using the scaling factors presented in Table 3.

- Mass ratio: $R_m = \frac{76}{M_S}$ (76 kg is the mass of the 50th percentile male),
- Stiffness ratio for pelvic tests: $R_k = \frac{32.9}{L_S}$ (32.9 cm is the pelvis width for the 50th percentile male),
- Stiffness ratio for thoracic tests: $R_k = \frac{30.4+34.9}{L_1+L_2}$ (30.4 and 34.9 cm are respectively the sub-sternal and axilary thoracic breath for a 50th percentile male),

Subject's acceleration were normalised using $R_a = \sqrt{\frac{R_k}{R_m}}$; time using $R_t = \sqrt{\frac{R_m}{R_k}}$ and impact force using $R_f = \sqrt{R_m \times R_k}$.

The corridors presented in this paper were calculated from the normalised responses with re-aligned peaks. The corridors were determined at each time step by computing the mean response and the standard deviation. The response corridor was created by plotting the mean value ± one standard deviation.

		For thoracic tests (LCC & MRL)					For pelvic tests (LCC & MRB)			
Subject	Mass [kg]	Sub-sternal width [cm]	Axiliary width [cm]	R_a	R_t	R_f	Bitrochanter width [cm]	R_a	R_t	R_f
SIBER 01	67	28	29	1.005	0.995	1.139	34.5	0.917	1.091	1.040
SIBER 02	33	25.3	25	0.751	1.332	1.729	30.5	0.684	1.461	1.576
SIBER 03	52	27	26	0.918	1.089	1.342	31	0.852	1.174	1.245
SIBER 04	50	23	28.5	0.913	1.095	1.388	31.5	0.829	1.206	1.259
SIBER 06	66.5	25.5	29	0.888	1.126	1.349	31.5	0.829	1.206	1.259
MRT 01	82						33	1.037	0.964	0.961
MRT 02	76						33	0.998	1.002	0.998
MRT 03	69						34	0.937	1.067	1.032
MRT 04	52						31	0.852	1.174	1.245
MRT 05	54	27.5	28.5	0.910	1.099	1.281	32	0.855	1.170	1.203
MRT 06	86	28.5	33	1.096	0.912	0.969	31	1.096	0.913	0.968
MRT 07	60	31	27	0.942	1.061	1.194	30	0.930	1.075	1.179
MRT 08	59.5	27	28.5	0.959	1.042	1.226	33	0.883	1.132	1.128
MRT 09	82	33	31	1.049	0.953	0.972	34	1.022	0.979	0.947
MRT 10	70	31.5	27	1.014	0.986	1.101	33	0.958	1.044	1.040

Table 3 : Scaling factors for thoracic and pelvic impact tests. LCC tests used SIBER subjects, MRL and MRB tests used MRT subjects.

RESULTS

AUTOPSY FINDINGS – In Table 4, AIS values for the thorax are given independently for the right and the left sides as they could have been caused by the left shoulder impact or the right thoracic impact. These AIS just give an overview of the severity of the tests but cannot be considered as a thoracic AIS.

Subject	Right fractured Ribs [nb of fx]	Thoracic AIS (Right)	Left fractured ribs	Thoracic AIS (Left)	Pelvic fracture	Pelvic AIS
SIBER 01	3	1	2,3[2], 4, 5	3	None	0
SIBER 02	5[2], 6[2], 7[2], 8[2], 9[2], 10[2]	4	1, 2, 3, 4, 5, 6	3	Right greater trochanter fx	Not coded according to AIS 90
SIBER 03	3[2], 4[2], 5[2], 6, 7, 8, 9	4	2, 3, 4, 5, 6, 7, 8	3	Right ilium branch fx	2
SIBER 04	2, 3, 4, 5, 6, 7, 8, 9[2], 10[2]	4	2, 3, 5	2	Right cotyle fx Right ilio and ischio-pubic branch fx	2
SIBER 06	2, 3, 4, 5, 6[2], 7[2], 8[2], 9	4	5	1	none	0

Table 4 : Summary of injuries sustained by SIBER subjects

During MRL tests on the thorax, few rib fractures occurred at low speed while serious thoracic fractures could be observed for higher impact speed. For the pelvic tests (MRB tests), no fracture was seen at 3.4 m/s, whereas 6.6 m/s tests led generally to AIS 2 injuries.

Due to rather high speed impacts on the thorax carried out without padding, the subjects sustained numerous rib fractures. However, the purpose of the present study is to analyse vertebral column kinematics and not to look at thorax behaviour under side impacts.

MEASUREMENTS – A first analysis was performed with only the SIBER experiments. MRL and MRB tests are added in the next section to define force-time and acceleration-time corridors at intermediate speeds. Furthermore, detailed kinematics analysis was not possible from these older tests as PMHSs' photographic instrumentation was simpler.

The impact force and acceleration measurements were filtered, normalised in value and time and re-sampled at 10 kHz. The start of the impact force responses was synchronized and the resultant acceleration peaks were aligned on the latest peak. Indeed, even if some curves were more comparable after the normalisation of the time, offsets between the responses still remained. As a consequence of this re-alignment, timing between acceleration measurements taken at the different vertebra levels could not be compared.

Thoracic impacts: Raw impact force responses are presented in Fig. 2 for comparison with the normalised and re-aligned responses. The normalisation largely increased the impact force response of the subject #2 (LCC03) because of its very low mass (33 kg). At this point, normalisation could be disputable when the subject anthropometry was too different from that of the 50th percentile. Therefore, tests conducted with subject#2 were not considered for impact force corridor definition (Fig. 3).

The average normalised impact force peak was equal to 2.1±0.2 kN (2.3±0.5 kN if subject #2 is considered) in comparison to 1.6±0.2 kN for raw data. The accelerations recorded at T12 level (close to T8, the point of impact) and sacrum level gave an idea of the thorax and pelvis coupling produced by the lumbar spine.

At T12 level, just below the impact location, a 20G resultant acceleration was recorded, whereas at the sacrum level, only 5G were transferred. The pelvis mass created a resistance to the pelvis motion, which was possible because of the lumbar spine flexibility. In Fig. 4 & Fig. 5, T12 and sacrum responses were re-aligned to make peaks occurred at the same time and defined corridors. In comparison to the lateral acceleration (a_y), a significant longitudinal acceleration (a_x) was also seen for three of the five subjects. Its peak was recorded between 10 and 20 ms. This could be due to the compression of the thorax which made the T12 vertebra first move rearward before rib fracture events.

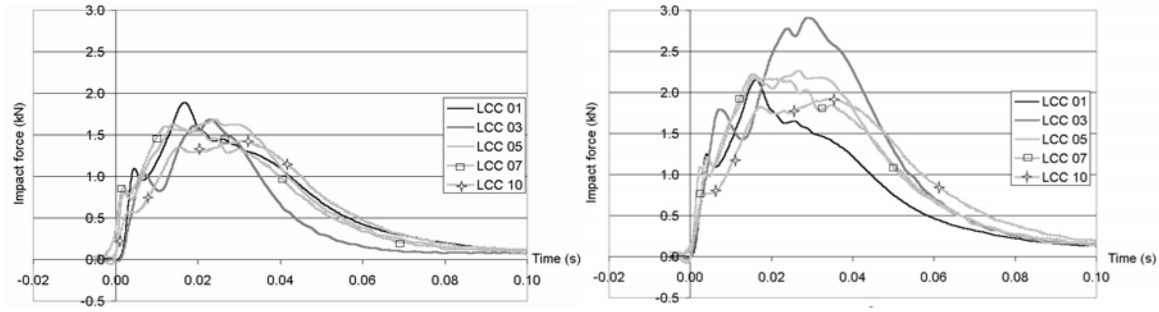


Fig. 2 : Impact force vs time for LCC thoracic impacts (v=4.1 m/s): Raw data (left) and Normalised / re-aligned data (right)

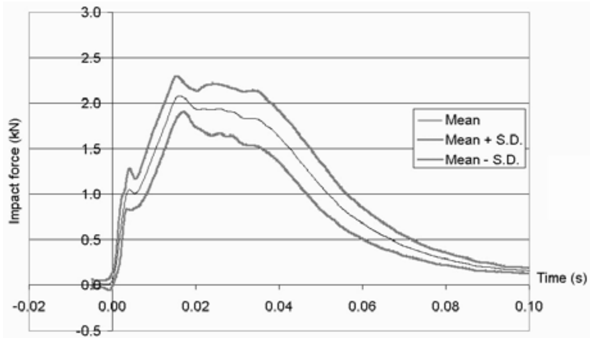


Fig. 3 : Impact force-time corridor for LCC thoracic impacts (V=4.1 m/s)

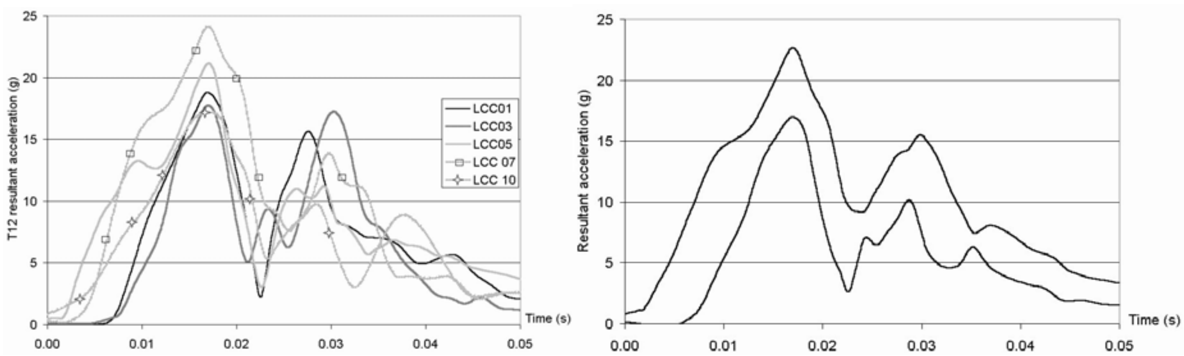


Fig. 4: T12 resultant accelerations: normalised and re-aligned (left)– derived corridor (right)

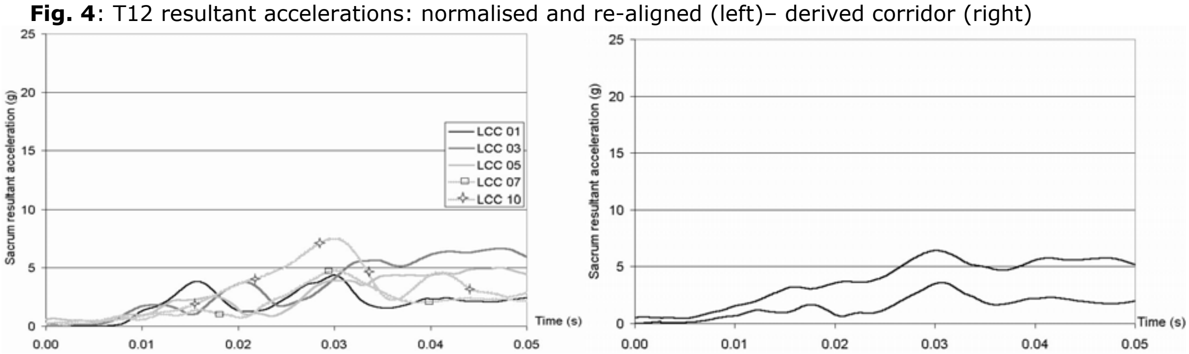


Fig. 5 : Sacrum accelerations: normalised and re-aligned (left) – derived corridor (right)

Previous tests (MRL tests) were conducted at two different speeds (3.3 and 5.5 m/s), this allows the definition of T12 resultant acceleration corridors above and below the 4.2 m/s one (see Fig. 6).

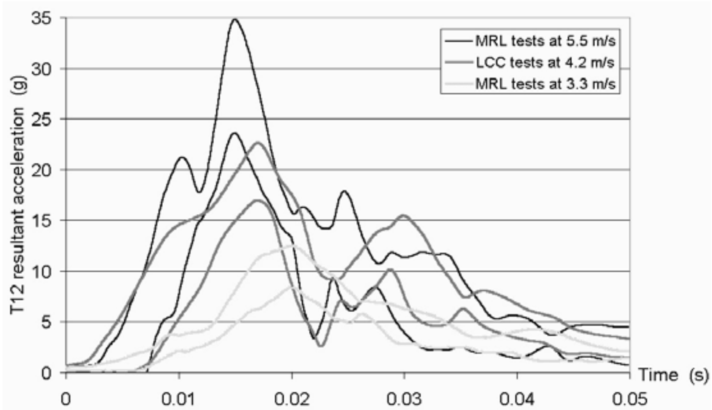


Fig. 6 : Superposition of T12 resultant acceleration corridors for 3.3, 4.2 and 5.5 m/s thoracic impacts (MRL and LCC test campaigns)

<u>Pelvic impacts,</u> The average normalised impact force peak was equal to 5.8±0.5 kN (6±0.7 kN if subject #2 is considered) in comparison to 4.7±0.2 kN for raw data (Fig. 7). Force-time response corridor drawn from all normalised responses is presented in Fig. 8. Response of subject #2 was excluded. The pelvic impacts were delivered at a higher speed (6.6 m/s) than the thoracic impacts resulting in higher T12 acceleration resultants. However, the created T12 accelerations, when the pelvis was impacted, were higher in proportion than those of the sacrum for thoracic impacts (thoracic impacts created a 17G T12 resultant acceleration and a 5G sacrum resultant acceleration, i.e. 70% reduction, whereas pelvic impacts created a 35G resultant acceleration and a 15G T12 resultant acceleration, i.e. 60% reduction). Considering non re-aligned responses in Fig. 9 and Fig. 10, T12 acceleration peak occurred 9 ms after that of the sacrum, except for subject # 2 which was immediately entirely accelerated due to its low mass.
For the MRB tests conducted previously, sacrum and T12 resultant acceleration corridors are also defined (Fig. 11 & Fig. 12).

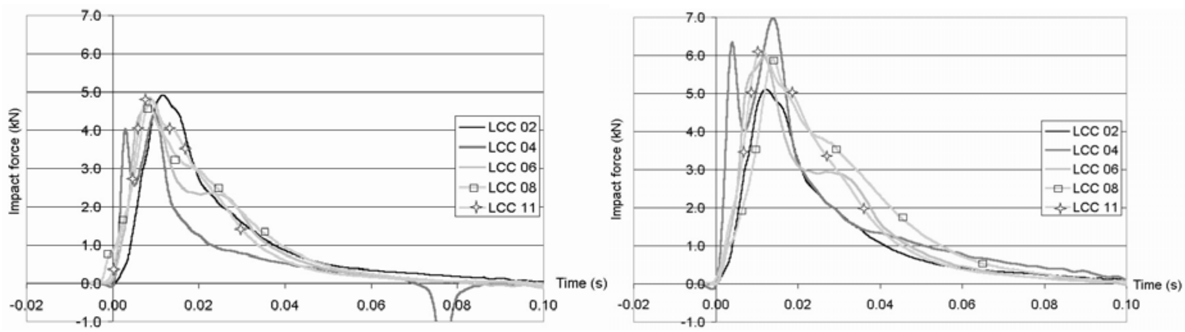


Fig. 7: Impact force for LCC pelvic impacts (V=6.6 m/s): Raw data(left) Normalised / re-aligned data (right)

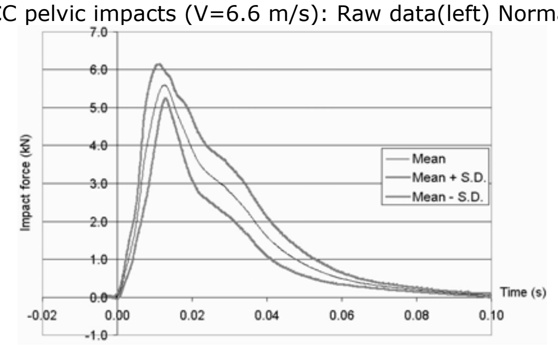


Fig. 8: Impact force-time corridor for LCC pelvic impacts (V=6.6 m/s)

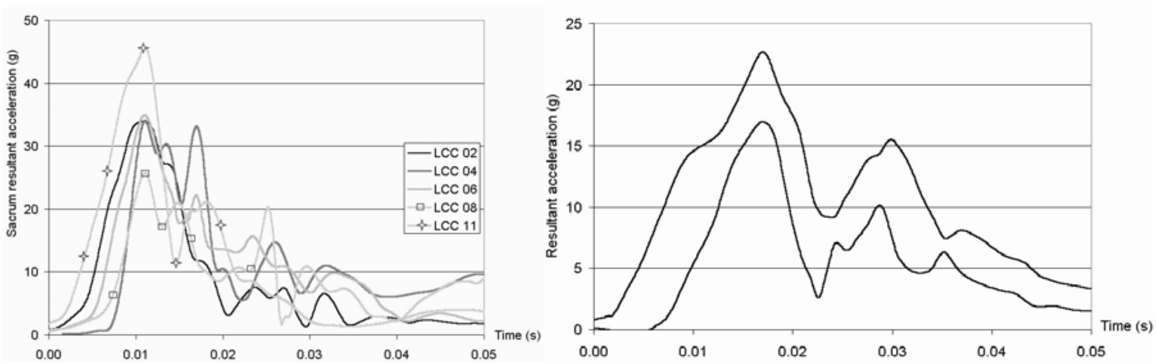


Fig. 9: Sacrum acceleration for LCC pelvic impacts (V=6.6m/s):normalised and re-aligned data (left) – derived corridor (right)

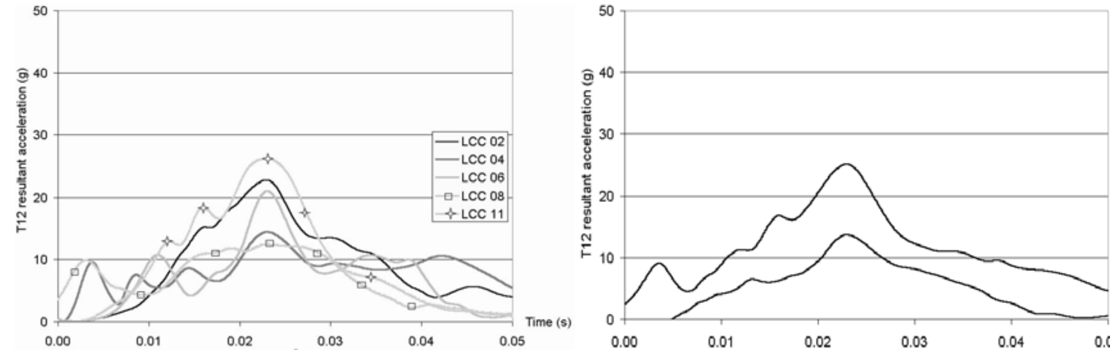


Fig. 10: T12 resultant acceleration for LCC pelvic impacts (V=6.6 m/s): normalised and re-aligned data (left)– derived corridor (right)

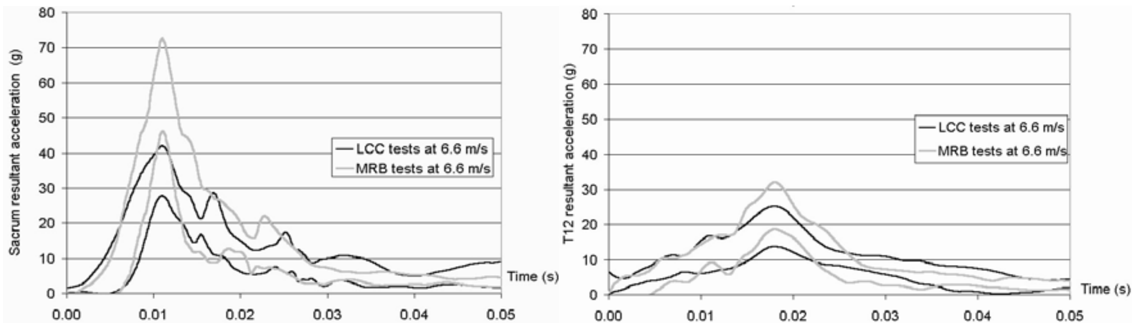


Fig. 11: Resultant acceleration corridors for 6.6 m/s MRB pelvic impacts: Sacrum (left) – T12 (right)

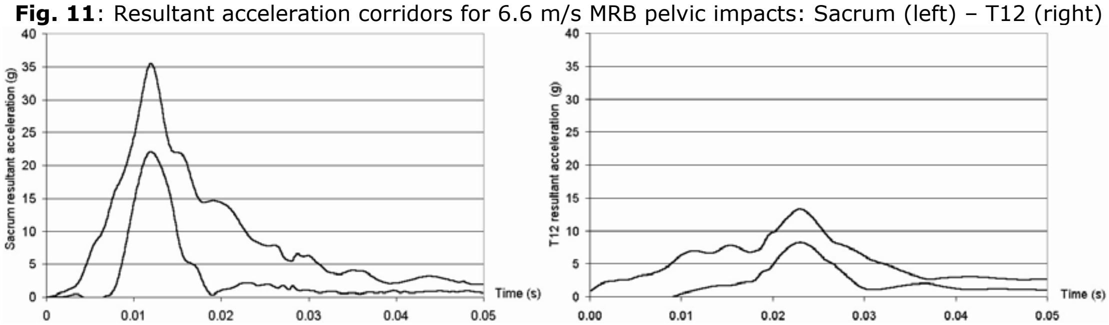


Fig. 12: Resultant acceleration corridors for 3.4 m/s MRB pelvic impacts: sacrum (left) – T12 (right)

KINEMATICS – The subjects' kinematics analysis was based on rear views of the subject. The global spine motion was deduced from the trajectories of the markers placed on the PMHSs. Every 10 ms, a line was drawn between the mid-point of the marker couples placed at T1, T4, T8, T12 and sacrum. The first line was drawn at 0 ms, the last one at 90 ms. During this 90 ms period, the consistency of fixed distance values was checked: for T12 and sacrum couple of markers, a maximum distance reduction of 6% was seen using the 2D film analysis.

The spine kinematics could be described through the change of angles calculated at different levels of the spine (Fig. 13). For instance, between T12 and the sacrum, the first $\alpha_{(T12\text{-sacrum})}$ angle represented the orientation of T12 with respect to the sacrum. The second $\beta_{(T12\text{-sacrum})}$ angle represented the orientation of the line drawn between the mid-points of T12 and sacrum couple of markers with respect to a vertical reference. Both angles were set to zero at the beginning of the impact; only relative angle variations were studied.

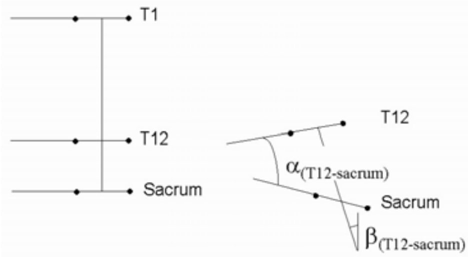


Fig. 13 : Definition of angles α and β

<u>Thoracic impacts</u>: in some tests, the PMHS's markers could not be tracked for some pictures (e.g. T8 and sacrum markers respectively LCC 02 and LCC 11 in the case of a pelvic impact). In test LCC 05 (thoracic impact), the PMHS had a different motion (T1 point didn't go down and sacrum point went upwards) because the electromagnet did not release the subject. This could be seen from the T1 point which stayed at the same height along the impact. Only one lumbar marker was placed on the first three subjects. For the two last subjects (tests LCC 07 & LCC 10), it was possible to fix several markers on the lumbar spine (respectively 5 and 4 markers).

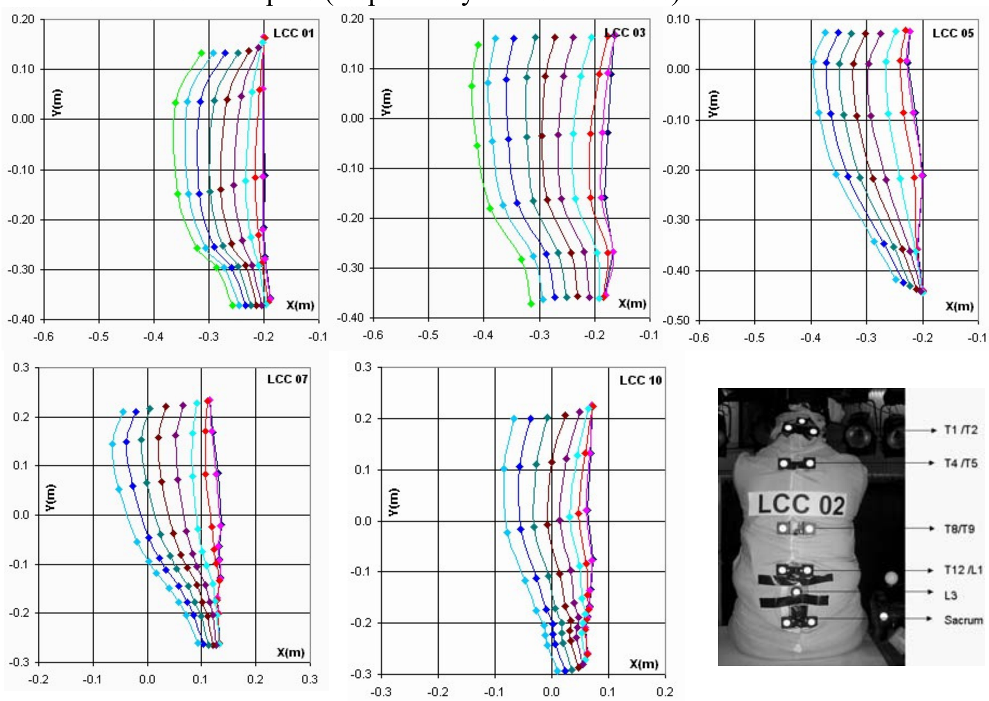


Fig. 14 : PMHSs' spine motion for thoracic impacts delivered at T8 level (Rear view)

Fig. 14 shows that the thoracic spine was strongly bended. The upper vertebrae translated laterally by approximately 150 mm whereas the pelvis motion was between 50 to 100 mm (Table 5). During the first 30 ms, T12 vertebra translated by 30 mm without any motion of the sacrum.

Test	Sacrum lateral motion [mm]	T12 lateral motion [mm]	T12-Sacrum relative motion [mm]
LCC 01	77	155	78
LCC 03	133	211	78
LCC 05	50	155	105
LCC 07	55	155	100
LCC 10	67	122	55
Mean	*77*	*160*	*83*

Table 5 : Comparison of sacrum and T12 vertebrae lateral motion during a thoracic impact

The curves of α and β in Fig. 15 and Fig. 16 confirm the bending of the thoracic spine (average $\alpha_{T1/T12}$ value of -25°) and the translation motion of the thoracic part ($\alpha_{T12/Sacrum}$ variation remains under 10°). Accordingly, the largest β variation was seen between T12 and sacrum level while it was low between T1 and T12. The smaller and later displacement of the pelvis, due to its mass induced a S-shape deformation of the lumbar spine.

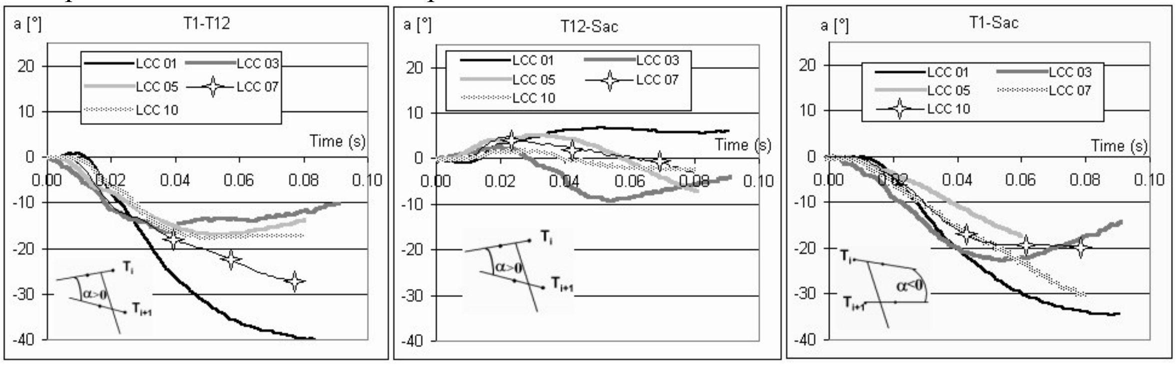


Fig. 15 : α-angles between T1-T12, T12-sacrum and T1-sacrum for thoracic impacts

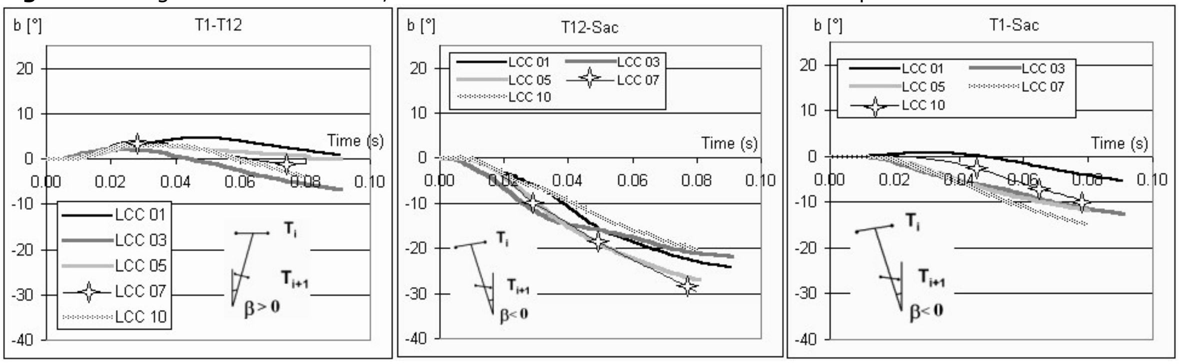
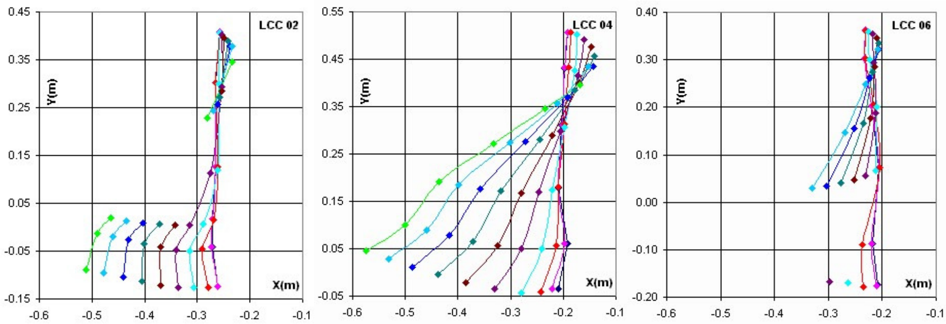


Fig. 16 : β-angles between T1-T12, T12-sacrum and T1-sacrum for thoracic impacts

Pelvic impacts, the subject rotated around an antero-posterior axis at around T4 level. The pelvis was pushed laterally between 200 to 300 mm and the motion of T12 was limited to 150 mm (Table 6). During the first 40 ms, the sacrum motion reached 80 mm while T12 stayed almost still (Fig. 17).

Test	Sacrum lateral motion [mm]	T12 lateral motion [mm]	T12/Sacrum relative motion [mm]
LCC 02	256	194	62
LCC 04	367	228	139
LCC 08	200	94	106
LCC 11	267	100	167
Mean	*273*	*154*	*119*

Table 6 : Comparison of sacrum and T12 lateral motion during LCC pelvic impacts



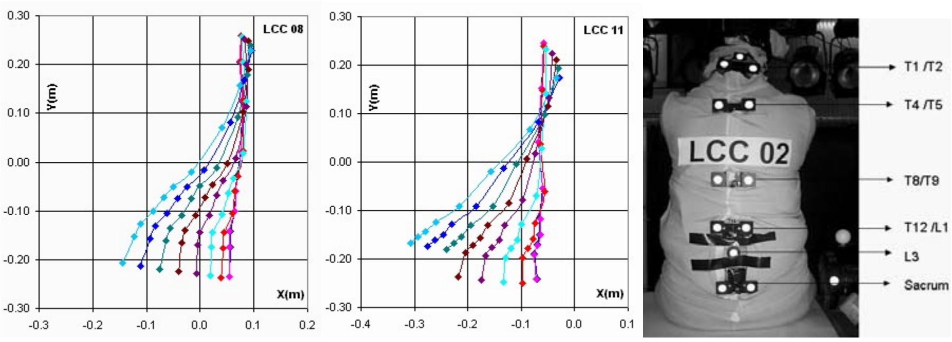


Fig. 17 : PMHSs' spine motion for pelvic impacts at greater trochanter level (Rear view)

The variation of $\alpha_{T1/sacrum}$ was under 10°, indicating that T1 and the sacrum remained parallel during the impact. The opposite variation of $\alpha_{T1/T12}$ and $\alpha_{T12/sacrum}$ during the first 40 ms showed the S-shape deformation of the lumbar spine.

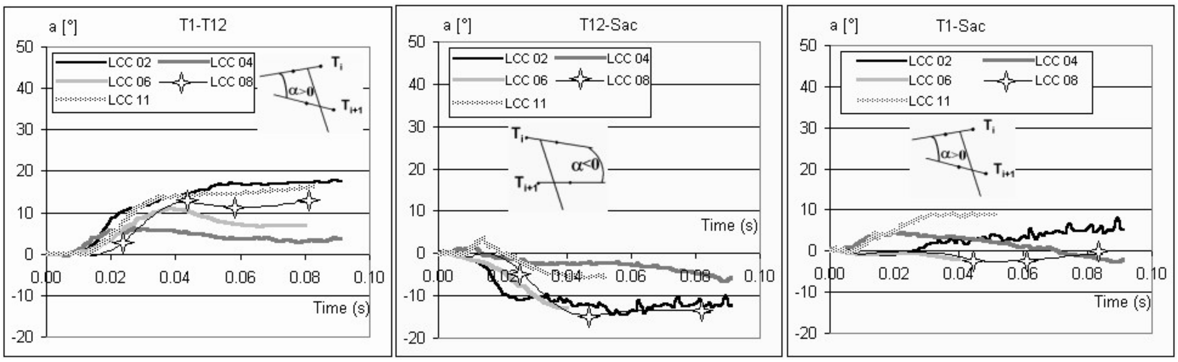


Fig. 18 : α-angles between T1-T12, T12-sacrum and T1-sacrum for pelvic impacts

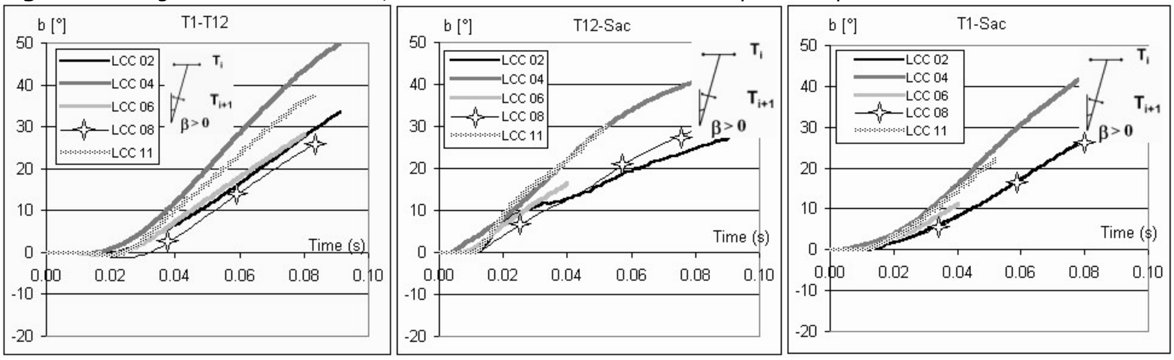


Fig. 19 : β-angles between T1-T12, T12-sacrum and T1-sacrum for pelvic impacts

EVALUATION OF THE WORLDSID DUMMY

The same test conditions were applied to the WorldSID prototype dummy (tests conducted at INRETS), the WorldSID pre-production dummy (tests conducted at TNO) and the ES-2 to compare their responses to that of the human subjects (Table 7 & Table 8).

Test	Dummy	Speed [m/s]	Impacted area	Configuration
LWC01 / 02	Prototype	4.13	Thorax wo arm	Dummy seated upright + padding
LWC03 / 04	Prototype	3.34	Pelvis	Dummy seated upright
LWC05 / 06	Prototype	6.87	Pelvis	Dummy seated upright
TNO-10599	Pre-production	6.63	Pelvis	Dummy in automotive posture
TNO-10648	Pre-production	6.63	Pelvis	Dummy seated upright + suit
TNO-10649	Pre-production	6.63	Pelvis	Dummy seated upright + suit
TNO-10656	Pre-production	6.68	Pelvis	Dummy seated upright + suit
TNO-10657	Pre-production	3.39	Pelvis	Dummy seated upright + suit
TNO-10658	Pre-production	3.39	Pelvis	Dummy seated upright + suit
TNO-10636	Pre-production	4.3	Thorax wo arm	Dummy in automotive posture
TNO-10638	Pre-production	4.3	Thorax wo arm	Dummy in automotive posture

Table 7 : WorldSID test matrix

When comparing the dummy response with that of the PMHSs, design limitations of the dummy shouldn't be forgotten. Indeed, the majority of the WORLDSID spine is rigid in order to accommodate data acquisition systems, and its lumbar spine is 150 mm shorter than the human spine (Table 8). This implicates a rather soft structure for the lumbar spine to compensate its short length and the rigidity of the thoracic spine. In the case of the ES-2, the spine dimensions are closer to that of a human.

Test	PMHS	T1-Sacrum length [mm]	T12-Sacrum length [mm]
LCC 01-02	SIBER 01	521-532	140-141
LCC 03-04	SIBER 02	523-541	198-214
LCC 05-06	SIBER 03	517-538	229-247
LCC 07-08	SIBER 04	495-494	241-256
LCC 10-11	SIBER 06	485-487	185-187
	Mean	*513 ± 21*	*204 ± 42*
LWC01-06	WORLDSID	403 ± 4	52 ± 2
ES-202	ES-2	530 (from CAD)	140 (from CAD)

Table 8 : Subject and dummy spine lengths (measured from film analysis if not specified)

Both WorldSID prototype and pre-production versions were compared with the PMHS corridors defined previously. In terms of resultant accelerations recorded at T12 and sacrum levels, the WorldSID dummy response peaks were of the same magnitude as for the PMHSs (Fig. 20 & Fig. 21). This indicated a rather good load path from the impact point to the lumbar spine area and also a human like coupling of the upper torso with the lower body. The biggest difference was seen for the T12 response in the case of a thoracic impact: the dummy response showed three peaks whereas the PMHSs' corridor gave a unique peak slightly above 20G (Fig. 20). This may be attributed to the rib or rib padding behaviour of the dummy. T12 and sacrum resultant acceleration obtained for the 3.4 m/s impacts (Fig. 22) were below the corridors but T12 deviation was proportional to the deviation of the sacrum acceleration. The reason for this was most probably due to the too low sacrum response and not necessarily to a lumbar spine deficiency.

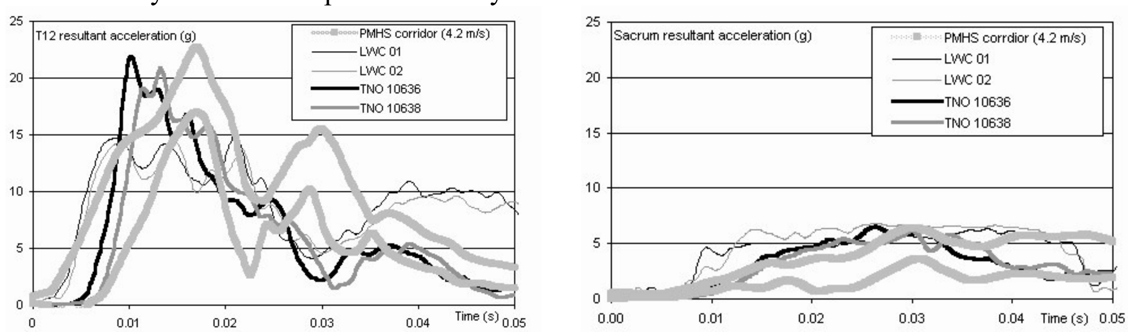


Fig. 20: WorldSID T12 and sacrum resultant accelerations with respect to PMHS corridors (bold grey lines) in the case of a thoracic impact (V≈4.2 m/s)

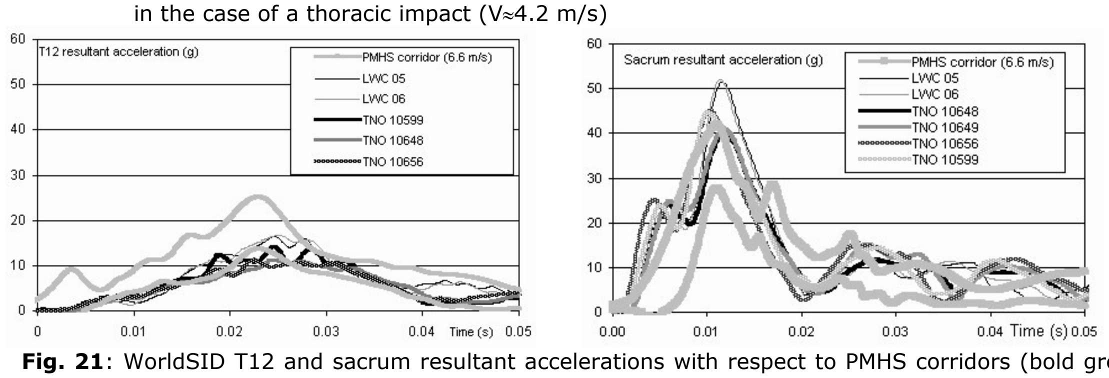


Fig. 21: WorldSID T12 and sacrum resultant accelerations with respect to PMHS corridors (bold grey lines) in the case of a pelvic impact at 6.6 m/s

The main difference between WorldSID prototype and pre-production version was the modification of the pelvis design which was softened in the WorldSID pre-production dummy. This explains why lower sacrum accelerations were recorded for this dummy (Fig. 21). Except for the shoulder rib, no modification was made to the thorax. Contrary to the prototype dummy, thoracic impacts on the pre-production dummy were carried out with the dummy suit. In addition to this latter difference, dummy initial position may explain differences between the two dummies seen on the T12 acceleration resultants (Fig. 20).

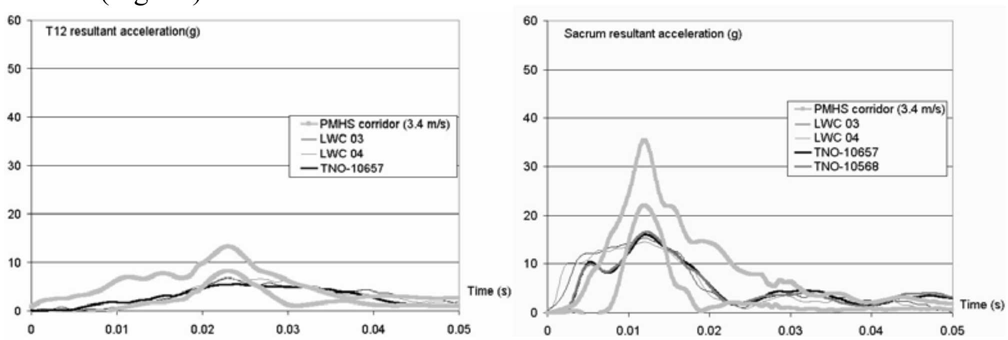


Fig. 22: WorldSID T12 and sacrum resultant accelerations with respect to PMHS corridors (bold grey lines) in the case of a pelvic impact at 3.4 m/s

Since the WorldSID dummy spine box was rigid from T1 to T12, markers were placed on it at T1 level, optionally at a middle level called T6, and two others were placed at the end of two pins extended outside the pelvis foam to allow tracking during tests. From these markers, trajectories of fictive T12 and sacrum points were calculated. The T12 position was deduced from markers positioned at the middle of the rigid spine (markers 5 & 6), point on sacrum from markers placed on sacrum block (markers 7 & 8). These two points corresponded to the projection on a vertical plane perpendicular to the camera axis of the central point of the upper surface of the lumbar spine and the mid point of the two fixations of the lower spine on the sacrum block. The coordinates of these two points were calculated from graphic construction taking into account part dimensions and positions measured before the impact. WorldSID and ES-2 dummies before the impact and their lumbar spine design are presented on Fig. 23.

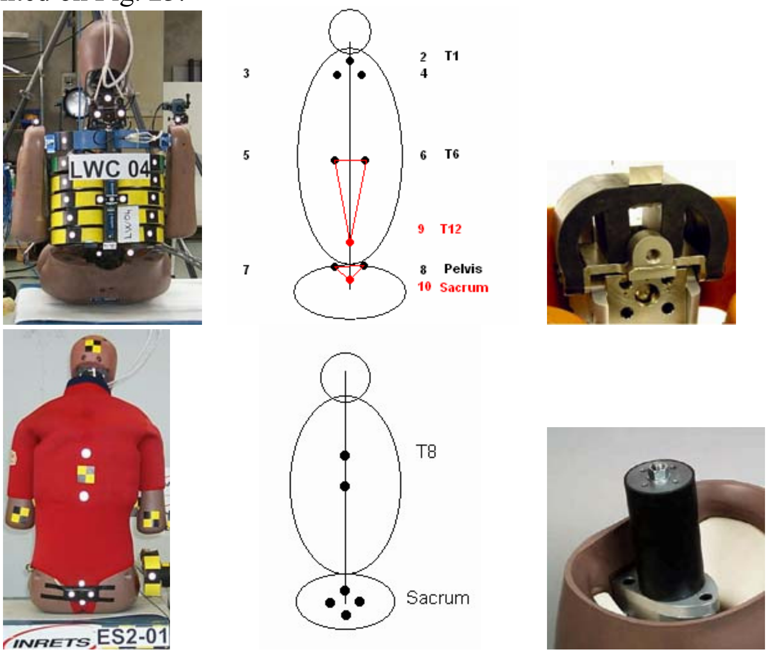


Fig. 23 : Photographic markers placed on the dummies and lumbar spine designs (top-WorldSID prototype and bottom-ES-2)

Dummy global kinematics of the spine and angle variations were studied similarly to the PMHSs. For the 4 m/s thoracic impacts, the thoracic spine of the WorldSID translated as a single piece. To compensate for the whole spine deformation, the dummy lumbar spine was highly deformed. The pelvis motion was higher and occured earlier as compared to the human. In the case of the 6.6 m/s pelvic impacts, the WorldSID pelvis moved 25 mm during the first 20 ms without any motion of the thorax. The spine then started to rotate around a point at T4 level, highly deforming the lumbar spine.

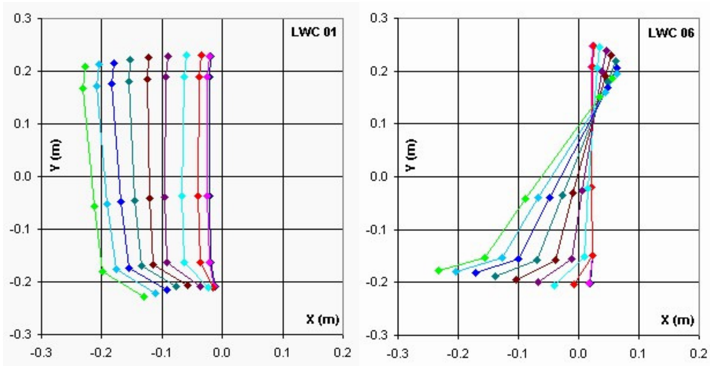


Fig. 24 : WorldSID spine kinematics under thoracic impact (left) and pelvic impact (right)

As dummy T1 and T12 belonged to the same rigid body, only angle variations between T1 and sacrum or T12 and sacrum were relevant. Furthermore, it was not possible to compare directly the dummy and PMHS spine orientation (angle β) because of their difference in length. One must note the differences in marker positions: PMHSs and ES-2 sacrum markers were at mid-height of the sacrum, whereas the point calculated on WorldSID lay upon the sacrum block, at the interface between the lumbar spine and the sacrum. Therefore, the WorldSID spine length was underestimated whereas that of the subjects was generally overestimated due to the necessity to fix markers.

T1-sacrum angles (Fig. 25- a & c) –The α angle reached high values (20 to 35°) depending on the subject whereas it remained low for the WorldSID. This was due to its rigid thoracic spine which made T1 to only translate. The β angle remainsed low (close to 5°) until 40 ms and the WorldSID behaviour is in line with the subjects' responses.

T12-sacrum angles (Fig. 25- b & d) – The α angle of the PMHS increased rapidly while that of the WorldSID stayed close to zero until 20 ms before going negative. The PMHSs' spine was bending while the dummy thorax started to translate. The flexibility of the WorldSID lumbar spine caused the β angle to increase through to 50 °, but this value could not be directly compared with that of the PMHS because of the different lengths.

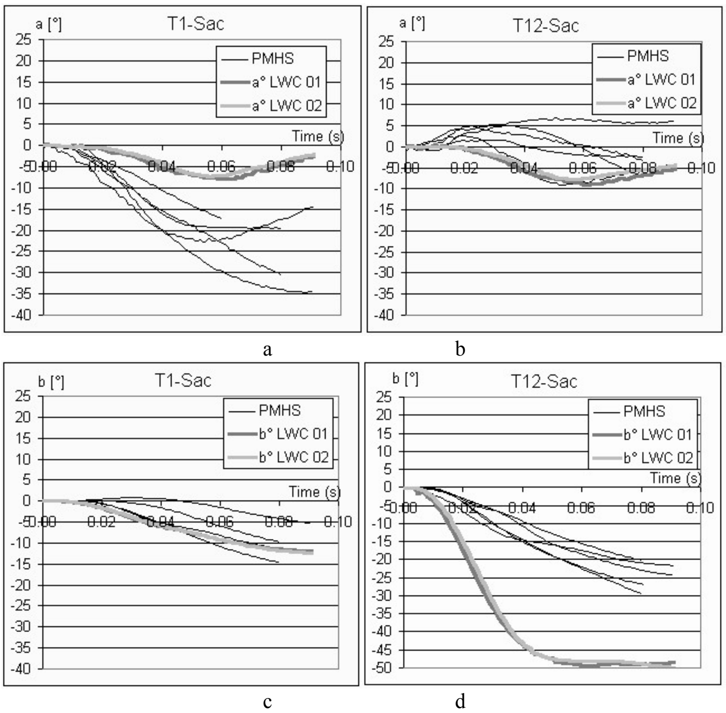
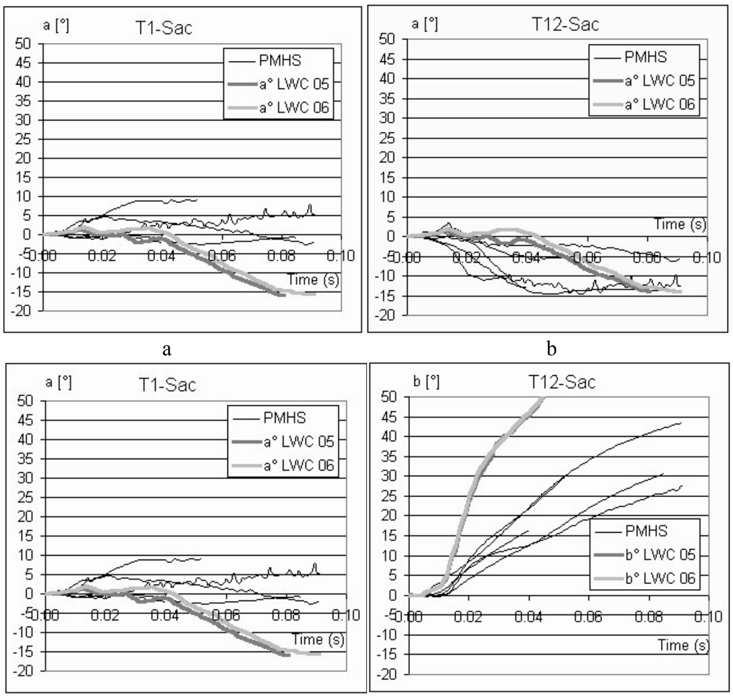


Fig. 25 : α and β angles in the case of a thoracic impact – Comparison between PMHS and WorldSID (LWC tests)

T1-sacrum angles (Fig. 26- a & c) – The α angle was very low (inferior to 5°) until 70 ms for the PMHSs and 50 ms for the WorldSID dummy. This indicated that for a pelvic impact, the sacrum stayed parallel to the T1. The increase of the β angle showed a translation of the pelvis with respect to T1 during 40 ms. During the entire impact, the β angle of the WorldSID dummy remained within the PMHS response corridors. In these test conditions, the dummy behaviour was close to that of the PMHS during the first 40 ms.

T12-sacrum angles (Fig. 26- b & d) – At the beginning of the pelvic impact, the WorldSID thorax did not move. The α angle was almost zero until 40 ms due to the translation of the pelvis with respect to the thorax. For two subjects out of three, the α angle took rather rapidly a positive value of 10 °, but the thorax did not move. As for the thoracic impacts, the β angle reached a high value because of the high flexibility of the WorldSID lumbar spine and the high constraints in this region. Looking at T12-sacrum level, the WorldSID lumbar spine behaved differently to that of the PMHSs.



a b

c d

Fig. 26 : α and β angles in the case of a pelvic impact – Comparison between PMHS and WorldSID (LWC tests)

For a better understanding of the different behaviours of the subjects and dummies, kinematics of their spine and orientation of T1 and sacrum are shown in Fig. 27 & Fig. 28 & Fig. 29. For pelvic impacts, the comparison with the ES-2 was also possible as similar tests were performed with it.

When impacting the thorax, much more translation was seen on the WorldSID dummy as compared to the PMHS. For the PMHS, the motion of the spine was more localised at the impact level, whereas the whole thoracic spine moved in the dummy, forcing the pelvis to translate earlier and to a larger extent. The translation of the thorax with respect to the pelvis for this subject and the WorldSID dummy was respectively 100 and 56 mm after 80 ms.

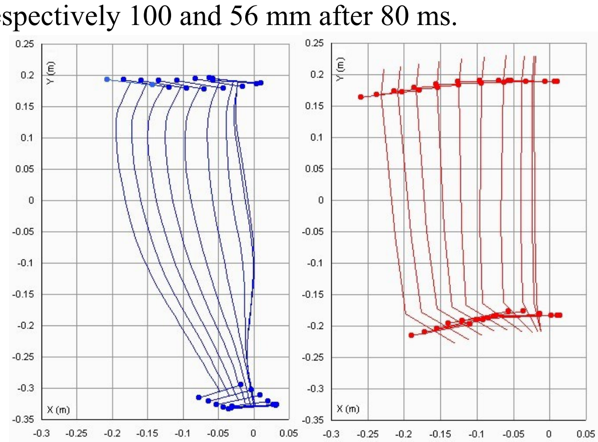


Fig. 27 : PMHS and WORLDSID spine kinematics for thoracic impacts at 4 m/s

In the case of the pelvic impact, a relative motion of the pelvis with respect to the thorax was seen for both the PMHS and the WorldSID dummy. On the contrary, for the ES-2 dummy, motion of the thorax was strongly coupled with that of the pelvis (Fig. 29).

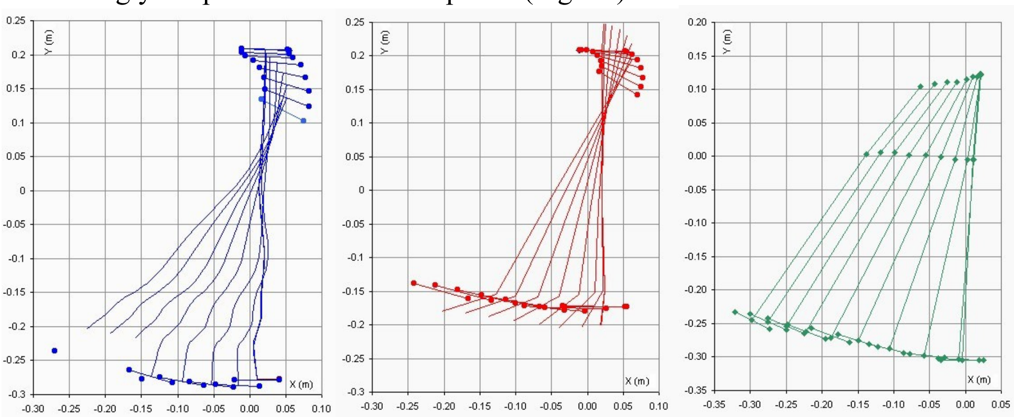


Fig. 28: Spine kinematics for pelvic impacts at 6.6 m/s: PMHS (left), WORLDSID (middle) and ES-2 (right)

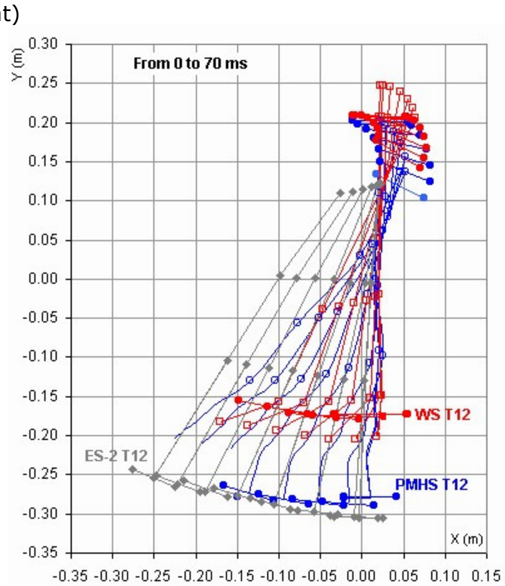


Fig. 29 : Superimposition of dummy and PMHS spine kinematics under pelvic impacts

DISCUSSION

In this study, impactor tests were carried out on five PMHSs in order to characterize the coupling between the thorax and the pelvis in lateral impacts. This analysis was based on T12 and sacrum accelerations and on 2D spine kinematics. The main limitation of this study was due to the use of PMHSs for which muscle tone did not exist. The muscles in living humans allow them to sit upright without any support and thus certainly have some influence on the thorax-pelvis coupling. In addition, to derive mean responses, data were normalised to reduce their scattering related to subject anthropometry variability. When the subject anthropometry was very far from the 50^{th} percentile subject, the normalisation was disputable and led to scattering the data instead of making them closer. The high ages of PMHSs, leading to low bone mineral density, were less problematic as injury threshold was not searched for.

The WorldSID dummy (prototype and pre-production versions) was evaluated against PMHSs' response corridors defined for resultant accelerations and kinematics. A comparison was made with the ES-2 kinematics during pelvic impacts.

The first difference between the WorldSID dummy and the human subjects was its lumbar spine length which was approximately 65 mm (from CAD) whereas that of the human was 150 mm (UMTRI, 1983). In addition, the whole human vertebral column was flexible whereas the flexibility of the WorldSID was concentrated at the lumbar level. Because of this design limitation, the dummy showed a different behaviour of its lumbar region to that of the human. The bending deformation seen on the PMHS thoracic spine could not be reproduced in the dummy which had a rigid thoracic spine. Therefore injury criteria based on dummy lumbar spine load measurements would be difficult to establish from human as WorldSID dummy and human lumbar spines have very different localised behaviour with a different functionality.

For thoracic and pelvic impacts, the whole spine orientation between T1 and sacrum level of the dummy was within the PMHSs' response corridors, whereas the angle between these vertebrae was almost equal to zero or below that of the human. This showed that more translation of the thorax with respect to the pelvis was seen on the dummy. However, the WorldSID lumbar spine response was globally acceptable because comparable resultant accelerations were found at T12 and sacrum levels between dummy and PMHSs, as well as similar overall kinematics. The dummy lumbar spine was shorter and softer than the human one to allow a comparable global spine response: in the case of a thoracic impact, since the thoracic spine was rigid, mainly translation of the thorax with respect to the pelvis was seen, whereas for the human, there was also a rotation and a deformation of the thoracic spine. In the case of a pelvic impact, dummy and PMHS kinematics matched quite well. Furthermore, the first approach in a comparison between the ES-2 and the WorldSID seemed to show that the WorldSID lumbar spine was an improvement for a more human-like coupling between the upper and the lower torso.

CONCLUSION

The comparison of the kinematics of PMHSs and WorldSID dummy submitted to the same thoracic and pelvic impacts gave a quantified evaluation of the WorldSID lumbar spine behaviour. Despite some limitations due to design constraints, the WorldSID lumbar spine showed a better biofidelic behaviour than its predecessors. This investigation was conducted using only impactor tests; the coupling seen between the WorldSID's thorax and pelvis, should also be checked in realistic side impact car crash conditions in order to verify that it still creates a human-like loading of the dummy by the door panel.

ACKNOWLEGMENTS

The authors would like to thank the INRETS/LBMC staff for their valuable contribution to this work: A. Maupas, co-ordinator of experiments, P. Lapelerie for preparing PMHSs, G. Goutelle, J. Lardière, M. Callejon for collecting films and photos, M. Marais, S. Serindat and JL. Russo for measurements taken during the tests.

The authors thank the department of Anatomy of Lyon-Nord (Medical University of Lyon), the Radiology laboratory of the Sainte Marie-Thérèse Clinic, Bron.

Special thanks to R. de Lange and B. Been (TNO Automotive) who performed the tests on the WorldSID pre-production version. The SIBER project, co-ordinated by the TNO and involving ACEA, BAST, FTSS-UK, INRETS, MIRA, TRL, TRW, WUT, is funded by the European Commission under contract G3RD-2000-00365.

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