CADAVER AND DUMMY INVESTIGATION OF INJURY RISK WITH ANTI-SLIDING SYSTEM IN CASE OF STATIC DEPLOYMENT

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ABSTRACT

In frontal impact, thorax and head injuries have strongly decreased with the development of occupant restraining systems including airbags, belt load limiters, and pretensioning systems. Nevertheless, the protection of abdomen and lower limbs has still to be improved, especially in rear seats. Indeed, car stiffness has increased in order to keep enough survival space for severe crashes. Thus, car manufacturers have developed specific restraint devices to improve protection of pelvis and lumbar spine, with prevention of submarining. One of these consists of an anti-sliding system based on an inflatable metallic wrap.

The main goal of the study was to investigate the risk of injury with a prototype of such a device, in case of static deployment, for in-position and out-of-position situations. Because the lack of relevance of the dummies in such conditions is suspected, and because criteria do not currently exist as far as the lumbar spine is concerned, six cadaver tests, including three in out-of-position situation, were carried out and duplicated with small female, and 50th male HIII dummies. Two inflators were used.

Cadavers were instrumented with linear accelerometers and angular velocity sensors for vertebra L2, L3, L5 and sacrum. The seat was equipped with load cells.

For the six cadaver tests, no injury was observed. Intervertebral rotation values are given for the cadavers and lumbar spine forces and moments recorded on dummies are presented. Comparisons regarding lumbar spine kinematics are realized for biofidelity assessment.

INTRODUCTION

In the last decade, the safety of car occupants involved in frontal crashes has been drastically improved thanks to the development of several new restraint technologies. The body areas where the changes were the most noticeable were clearly the head, the thorax and the lower extremities. As a consequence, issues such as submarining became of higher relative importance.

In order to avoid submarining while maintaining the comfort features of the seat, several active "anti-sliding" devices were developed by suppliers. Most of them were based on an inflator-propelled obstacle initially located under the seat cushion. These devices were designed to be ignited as soon as the crash is detected such that an obstacle moves toward the buttocks in order to prevent penetration motion of the pelvis into the seat cushion. As a result, the pelvis is coupled to the seat very early in the crash time history and its rotation is locked such that submarining is prevented.

OBJECTIVES

The first objective of our study was to investigate the potential intrinsic aggressiveness of such a concept on the pelvis and lumbar area for both In-Position (IP) and Out-of-Position (OOP) situations, in case of static deployment. The second objective was to compare the lumbar spine and pelvis kinematics of the dummies and the cadavers in order to assess the relevance of the dummies in these specific situations.

LITERATURE REVIEW

Several studies dealing with the axial compression resistance of either isolated human cadaver vertebrae or vertebral units were available. A few studies on whole cadavers were also available. The tests conditions and results of these studies are presented in the table A1 of the annex.

Isolated vertebrae resistance

The tests were performed in quasistatic. The vertebrae ranged from C1 to L5. In all the studies reported, the vertebrae samples were relatively large, however the results somehow varied from one study to the other.

Coltman [3] tested 75 vertebrae (T1 to L5) in pure compression up to 70% of axial deformation of the vertebrae. The rupture compression force was 5900 N. This value is in good agreement with several other studies. Yoganandan [11] tested 63 vertebrae and reported a rupture compression force equal to 4.6 kN for the lumbar spine (no further information about the exact vertebrae). Myers [8] tested 61 vertebrae (L2 to L4) and reported a rupture compression force equal to 5600 N. In the same conditions, Brinckmann [2], Hutton [6], Kazarian [7] and Yamada [14] reported forces ranging from 4.4 kN to 8.3 kN.

In another study performed on 530 vertebrae (C1 to L5), Gozulov [5] reported, however, a rupture compression force equal to 13 kN. This value was sensibly higher than those reported above.

Vertebral unit resistance

A vertebral unit is defined as two adjacent vertebrae and connective structures (including the disk) in between. Tests were conducted at several compression speeds on several types of vertebral units. In all the tests, the loading was a pure compression. Brinckmann [2], Yoganandan [11] and Myklebust [9] reported the same value for the rupture compression force : 5 kN (5.5 kN for Yoganandan), while Hutton [6] and Willen [13] reported higher values approximately equal to 11 kN.

Whole spine resistance

Myklebust [9] conducted tests on 4 whole cadavers where the spine was loaded in compression through a force applied on T1. The thorax was kept vertical while the neck was flexed such that it was horizontal. A 15 cm x 15 cm plate then pressed the neck in order to apply a vertical force on T1. The compression rate was 10 mm/s. Two plates were placed on each side of the thorax in order to avoid the lateral motion of the thorax. The skin was removed in front of the spine in order to allow a direct seeing of the vertebrae movements during the loading. For a compression force equal to 2.8 kN, crushing fractures were observed. The slope of the fractures ranged from 28° to 50°. For 3 specimens, fractures occurred between T10 and L2 while on the 4th specimen, it occurred on T7.

Synthesis of resistance

From the literature review, it appears that almost all the studies deal with the fracture vertebra mechanism by compression, except a few of them that deal with combined flexion and compression, which seems to be our case.

From these studies, one can find the following tendencies:

- The maximal compression force decrease when going up from the lumbar to the cervical spine (about 1 kN each 3 vertebrae)
- Dynamic loading at 100 mm/s increases the force rupture by 1.5 kN from static loading
- The maximal compression force decrease with age.

Gathering all these data and as a first approximation, the tolerances for pure compression and flexion-compression loading are summarized in the Table 1, which can be used as a reference for risk evaluation on human subjects.

Table 1. Tolerances for lumbar spine.

	Pure compression	Compression-flexion	
	Fz (kN)	Fz (kN)	My (Nm)
20-40 years	8	3	400
40-60 years	6.5	2.5	300
> 60 years	4	1.5	200

Comparison between human subject and dummy lumbar spines

Demetropoulos [4] has performed ten cadaver tests on complete isolated lumbar spine, without muscles. These tests were duplicated with HIII dummy. The results have shown that the HIII lumbar spine is stiffer than the human subject lumbar spine (ratio of 20/1 in flexion and 2/1 in extension). But the HIII lumbar spine represents the overall resistance in flexion including lumbar spine, muscles and abdomen. So, it is difficult to assess the real difference in angular stiffness between human subjects and dummies.

No criterion and no protection limit is currently available for the dummy lumbar spine. A under-evaluation of the risk of injury due to the poor relevance of dummies was feared in IP and OOP situations. As a consequence, tests were carried out on cadavers.

MATERIAL AND METHODS

Loading device

An “anti-sliding” device based on a metallic inflatable cushion was chosen for our study because prototypes were available. This device, prior to ignition, was located under the seat cushion at its forward portion (Figure 1).

Two types of inflators were tested: a standard one and a boosted one.

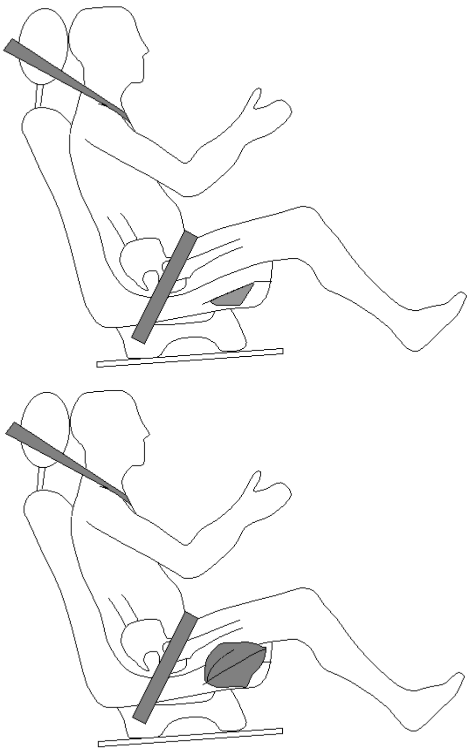

Figure 1. "Anti-sliding" device prior to activation (top view) and when activated.

Specimen

A total of six human cadavers were acquired for this study. There were two females and four males, ranging in age from 47 to 78. Specimens were obtained through the Body Donation Department of the Descartes University in Paris (France). The specimens were not embalmed to prevent undesirable changes in tissue properties. Table 2 shows the anthropometry and age for each cadaver.

Table 2. Specimen anthropometry.

Specimen #	Sex	Height (cm)	Age	Weight (kg)
544	M	169	70	82
545	F	166	64	64
547	M	179	78	70
548	F	163	47	55
546	M	166	67	50
549	M	164	74	58

Dummies

The tests on cadavers were duplicated with the 50th centile male and 5th centile female dummies.

Instrumentation

Seat

The seat was mounted on the test rig through four 3-axis load cells. A 2-axis accelerometer was fixed on the structure of the seat cushion.

Dummies

The HIII dummies were instrumented with head accelerations (x, y, z), upper neck forces and moment (Fx, Fz, My), thorax accelerations (x, y, z), one thorax angular velocity (ωy), lower lumbar spine efforts, moment and accelerations (Fx, Fz, My, Ax, Ay, Az), pelvis accelerations and angular velocity (Ax, Ay, Az, ωy), and femur forces, moments and accelerations (Fx, Fz, My, Mx, Ax, Az). In addition, for the HIII 50th percentile dummy, the upper lumber spine loads and moment were recorded. The SAE J211 recommended practice was used for filtering and sign convention.

Post mortem human subjects (PMHS)

The lumbar spine of the cadavers was instrumented at L2, L3, L5 vertebrae and on the sacrum using cubes equipped with 3-axis accelerometers and 1 MHD aligned along Y axis (Figure 2).

The femurs and tibias were instrumented with one 3-axis accelerometer each.

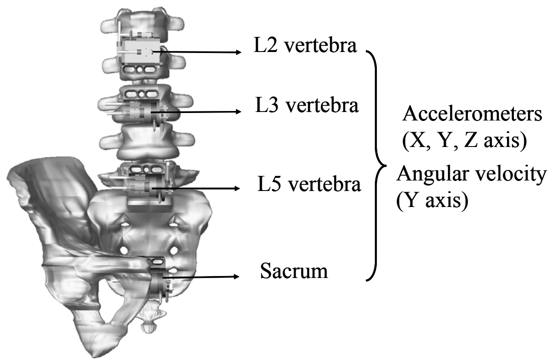

Figure 2. Instrumentation of the lumbar spine and sacrum area on cadavers.

Specimen initial position

Two positions were defined, one for IP and one for OOP situations.

The IP position (Figure 1) was a standard belted driving position (slope of the back seat = 25°, slope of the seat cushion = 5°). A foot rest was installed and adjusted such that the femur angle relative to the horizontal was 18°. A pretensioning system was installed at the buckle anchorage of the seat-belt. The time-to-fire of the anti-sliding device and the pretentioner were the same.

The OOP position corresponded to a passenger seating unbelted with both feet on the dashboard (Figure 3). The seat back slope was 45°. In such a position, the sacrum was exactly in front of the inflatable device (the ischiatic tuberosity was 150 mm backward from the fore edge of the cushion). The femur angle relative to the horizontal was 18°. This situation was assumed to be the worst case (i.e. with the higher pelvis and lumbar injury risk).

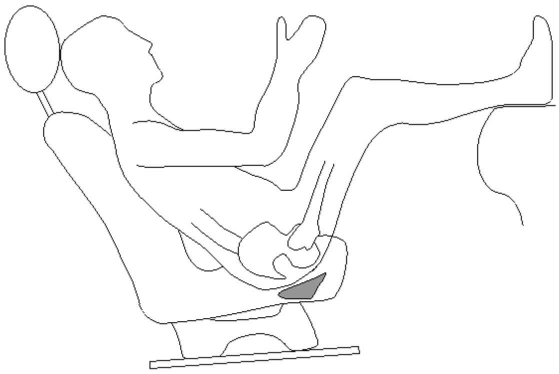

Figure 3: OOP position with the anti-sliding device prior to activation.

The initial position of the specimen was recorded through targets digitized using a Römer 3D arm. Additional attention was paid to the position of the spine cubes relative to the vertebrae landmarks.

Test matrix

The test matrix is presented in Table 3. Note that for all the tests performed in IP, the cadavers or dummies were belted except in test AA6-545 where the cadaver was unbelted.

Table 3. Test matrix

TEST CONDITIONS			HIII DUMMY TESTS		CADAVER TESTS	
Position	Inflator	Belt	Test number	Centile	Test number	PMHS
OOP	Standard	No	AA02	50th	AA07	547
OOP	Standard	No	AA04	5th	AA08	548
OOP	Boosted	No	AA12	50th	AA13	549
IP	Standard	Yes	AA01	50th	AA05	544
IP	Standard	Yes*	AA03	5th	AA06	545
IP	Boosted	Yes	AA11	50th	AA13	549

* Unbelted for cadaver test

RESULTS

Input loads

Figure 4 shows a typical time history of the loads applied on the pelvis with a boosted inflator. The force direction is roughly 55 degrees towards a horizontal axis.

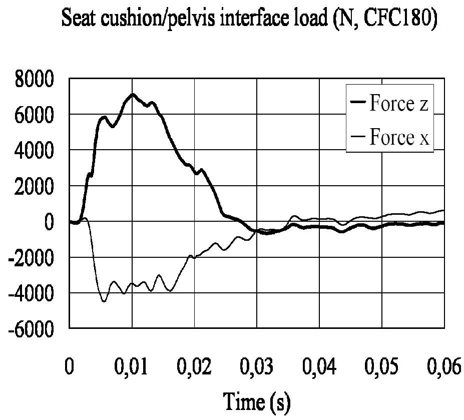

Figure 4. Time histories of seat cushion/pelvis interface forces

Dummy test results

Maximum dummy forces and moments on the lower lumbar spine, chest and pelvis resultant accelerations, thoracic and pelvic rotations are displayed in table 4 on the following page. Mean values, calculated from the three OOP and three IP tests, are also given.

The maximum resultant pelvic acceleration values are very close whatever the situation, with the highest values for the 5th percentile dummy. However, the film analysis shows noticeably different pelvic kinematics according to the situation. In IP, the pelvis moves back, because of the belt tension. In OOP, the pelvic movement according to the z axis is predominant.

On average, the thoracic resultant accelerations are quite half as high than the pelvic resultant accelerations.

The results do not show clearly the effect of inflator, boosted or not, indeed whatever the body part. On the other hand, the situation has an effect on the lumbar spine forces and moments. The compression forces are higher in OOP than in IP. The highest value (2.97 kN) is obtained with the 5th percentile dummy, in OOP situation.

Regarding the lumbar spine moments, flexion in IP is predominant. In OOP, during the first 100 ms, lumbar spine extension is observed, flexion appearing in a second phase. The results show negative pelvic rotations, in all the tests, with a magnitude lower than 6 degrees.

Although the input force seems to be high, the lumbar spine forces and moments are low. In OOP or in IP, inertia effect of pelvis and lower limbs mainly counterbalances the action force of the seat cushion.

Table 4. Main peak values for the dummy tests.

	Out-Of-Position			In-Position			OOP	IP
Test conditions	AA02	AA04	AA12	AA01	AA03	AA11	Mean values	
HIII dummy centile	50th	5th	50th	50th	5th	50th		
Inflator	Standard	Standard	Boosted	Standard	Standard	Boosted		
Pelvis								
Resultant acceleration (g)	21,3	26,1	24,1	22,9	26,9	21,3	23,8	23,7
Rotation (y, degrees)	-7	-8	-7	-5	-8	-6	-7,3	-6,3
Lumbar spine								
Momentum (Nm)	-65	-35	-56	117	80	102	-52	100
Fx (N)	1570	1100	2370	-680	-710	570	1680	-273
Fz (N)	-1970	-2970	-2370	-1240	-1230	-1190	-2437	-1220
Thorax								
Resultant acceleration (g)	10,9	17,6	14,6	7,1	13,7	10,7	14,4	10,5
Rotation (y, degrees)	-6	-5	-5	2	2	3	-5,3	2,3
Pelvis/Thorax								
Rotation (y, degrees)	-6	-5	-5	2	2	3	-5,3	2,3

Cadaver test results

Injury assessment

Autopsies were performed after the tests. No injury was observed on any cadaver.

Test analysis

Table 5 displays the main peak values obtained from the cadaver tests. In the last columns, mean values of the IP and OOP tests are given, with the available data. The intervertebral angular motions are calculated from the angular velocities. The intervertebral angular motion for the L5-L4 and L4-L3 units are estimates (L5-L3 angular motion divided by 2). The results show that the intervertebral angular motion is higher in OOP than in IP, with the most important values localized at the sacrum/L5 unit. The same observation can be done for the resultant accelerations. The differences between IP and OOP seem to be amplified by the 10 kg difference between the two groups.

Regarding the resultant acceleration values, differences between vertebrae and pelvis are not noticeable.

The results do not show a clear effect of the inflator. The highest values are obtained for the test AA09 with a boosted inflator, but also with the lightest cadaver.

In all the situations, the sign of intervertebral angular motions is always positive, indicating a flexion mechanism in IP or in OOP.

Table 5. Main peak values for the cadaver tests

	Out-Of-Position			In-Position			OOP	IP
Test	AA07	AA08	AA09	AA05	AA06	AA13	Mean values	
Subject number	547	548	546	544	545	549		
Subject mass (kg)	70	55	50	82	64	58	58,3	68,0
Inflator	Standard	Standard	Boosted	Standard	Standard	Boosted		
Acceleration (resultant, g)								
Sacrum	29	52	NA	19	NA	31	40	25
L5 vertebra	37	31	70	25	22	33	46	27
L3 vertebra	34	31	66	26	21	25	44	24
L2 vertebra	45	32	68	25	18	19	48	21
Angular motion (degrees)								
Sacrum	22	25	NA	9	20	7	23	12
L5 vertebra	16	16	20	5	7	4	17	6
L3 vertebra	11	13	18	4	2	-4	14	1
L2 vertebra	10	14	17	3	1	-4	14	0
Intervertebral rotation (degrees)								
Sacrum/L5	7	16	NA	4	13	3	12	7
L5/L4 and L4/L3 (estimated)	4	2	2	1	4	2	3	2
L3/L2	4	5	5	2	2	2	5	2
Sacrum/L2	20	24	NA	7	22	9	22	13

NA : not available

DISCUSSION

The first step in the evaluation of a safety system or its unwanted effects, consists in running tests using dummies and compare the criteria recorded on dummies to the tolerances on human being. In our study, although dummies are equipped to measure lumbar forces, their behaviour is questionable for the kind of loading caused by the anti-sliding device. As of today, no limits of tolerance are available for them specifically, and their poor biofidelity in this body area does not allow the direct application of human criteria and tolerances.

However, in spite of the force order of magnitude applied to the pelvis (up to 7 kN), lumbar forces measured on the dummy suggest that the injury risk associated to compression, flexion of both of them remains very low. Nevertheless, to confirm the harmlessness of the device and at the same opportunity to evaluate the dummy response and ability to assess the injury risk, tests were performed on PMHS. No injury was observed in the worst OOP case. It confirmed that the device is safe even with a boosted inflator whatever the specimen anthropometry.

However, the comparison of dummy and PMHS kinematics showed fundamentally different behaviors. The spine of the PMHS was always flexed while the dummy spine was mainly extended during OOP tests. Moreover, the compression of the lumbar spine was predominant for the dummies while the lumbar spine flexion seems to be the main mechanism for PMHS.

This difference of behaviour can be explained by the different initial positioning. The dummy remained straight even in OOP while the PMHS leaned in the seat. In addition to the geometrical differences the lumbar spine stiffnesses are significantly different between the cadaver and the dummy. Both differences highlight the poor ability of the dummy to reproduce realistic loading modes and consequently evaluate the injury risk.

This study did not provide means for lumbar spine characterization, especially since no forces were measured on PMHS. However, useful information are provided for the validation of a mathematical model of the human being, capable to mimic the kinematics of lumbar vertebrae.

CONCLUSIONS

Six dummy tests were performed with prototypes of a new concept of inflatable anti-sliding system. The test conditions included In-Position and Out-Of-Position situations, in order to evaluate the lumbar injury risk in case of static deployment. The forces and moments recorded on the dummy lumbar spine were very low and no risk of injury was suspected.

Nevertheless, six PMHS tests were also performed to complete this statement. The results confirmed that the device was safe. However, they also demonstrate that the dummy had not the same behaviour than the PMHS and by the way, was not able to assess properly the injury risk.

Research has then to be undertaken regarding the lumbar spine, where the protection criteria on dummy will become an issue to evaluate such protection systems.

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APPENDIX

Table A1: Literature review synthesis.

Author	Réf.	Specimen	Loading	Loading rate	Record	Injury
Brinckmann	[1] [2]	134 units of 2 vertebrae (T10 to L5, apophyses included)	Pure compression.	1 kN/s with a preload of 1 kN during 15 mm	F rupture = 5 kN Range from 4.4 to 6.8 kN	Fx of the vertebrae body
Coltman**	[3]	75 isolated vertebrae from T1 to L5	Pure compression up to 70% of the height.	89 mm/s (no preload)	F rupture thorax = 4.5 kN lumbar = 5.9 kN	Crush fx (occurs between 3% and 10% of deformation)
Gozulov	[5]	530 isolated vertebrae from C1 to L5	Pure compression.	Ranged from 0.08 and 1.7 mm/s	F rupture =13 kN	Depends on the deformation
Hutton	[6]	33 units of 2 vertebrae from L1 to S1, apophyses included	Pure compression.	3 kN/s with a preload of 1 kN during 5 mm	F rupture = 5.6 kN for L1, up to 8.3 kN for L5	Fx of the vertebrae body
Kazarian	[7]	61 isolated vertebra	Pure compression.	8900 mm/s 89 mm/s 0.89 mm/s	F rupture thorax = 6.8 kN	Crush fx
Myers	[8]	61 isolated vertebra : 22x L2 22x L3 and 17xL4	Pure compression.	1.5 mm/s (no preload)	F rupture =5.6 kN	Fx of the vertebrae body
Myklebust	[9]	14 whole spines T3 to L5	Pure compression (neck flexed)	Ranged from 10 to 1200 mm/s	F rupture 2.1 kN	Wedge crush fx
		13x T7-T12	Pure compression up to 50% of the height	1 mm/s.	F rupture = 3.3 kN	Fx of the vertebrae body
		9x L1-L5	Pure compression up to 50% of the height	1mm/s.	F rupture 5 kN	Fx of the vertebrae body
		4 whole cadavers	Force applied on T1	10 mm/s	F rupture =1.1 to 2.8kN	Wedge crush fx
Osvalder	[10]	16 units	Flexion - shearing	Static	Fx=0.62 kN My=160N.m	"flexion – distraction" type fx
Yoganandan	[11]	63 isolated vertebrae	Pure compression up to 50% of the height	2.54 mm/s	F rupture thorax = 3.3 kN lumbar= 4.6 kN	Vertebrae crushed
Yoganandan	[12]	38 isolated vertebrae	Pure compression	2.5 mm/s		Vertebrae crushed
		18 whole cadavers	Compression, flexed spine	2.5 mm/s	Fz comp = 2.5 kN associated with My = 170 Nm	Wedge crush fx
Willen	[13]	7 units of 3 vertebrae (T12-L2)	Pure compression.	Free fall of a 10 kg mass from 2 m.	F rupture 11 kN	"burst fracture" type fx