Chest injuries of elderly Post Mortem Human Surrogates (PMHS) under seat belt and airbag loading in frontal sled impacts. Comparison to matching THOR tests.

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ABSTRACT (word count: 395)

Objective

In 2020, according to the European Commission, 28% of Europeans will be aged 65 and older. Thus, protecting elderly car occupants is at the same time a priority and a challenge from the biomechanical point of view. The goal of the study was to develop experimental chest loading conditions that would cause up to AIS 2 chest injuries in elderly occupants in moderate-speed frontal crashes. The new set of experimental data was also intended to be used in the benchmark of existing thoracic injury criteria in lower-speed collision conditions.

Methods

Six male elderly (age>63) PMHS were exposed to a 35 km/h (nominal) frontal sled impact. The test fixture consisted of a rigid seat incorporating a ramp in the frontal direction, rigid footrest and a cable seat back. Two restraint conditions (A and B) were compared. Occupants were restrained by a force limited (2.5 kN (A) and 2 kN (B)) seat belt and a pre-inflated (16 kPa (A) and 11 kPa (B)) airbag (see picture). Condition B also incorporated increased seat friction. Matching sled tests were carried out with the THOR M dummy. IRTRACC readings were used to compute the predicted chest injury risk. Kinematics were recorded by a 3D motion capture system at 1 kHz. Three-axial acceleration and angular rate were measured at several locations along the spine, the head and the pelvis (10 kHz). PMHS were exposed to a post-test injury assessment. Tests were carried out in two stages, using the outcome of the first one to adjust the test conditions in the second. All procedures were approved by the relevant Ethics board.

Results

Despite the low load limiter and the use of the airbag, restraint condition A of three PMHS resulted in an unexpected high number of rib fractures (fx) (10 fx, 14 fx, 15 fx). Under condition B, the adjustment of the relative airbag/occupant position combined with a lower airbag pressure and lower seat belt load limit resulted in a reduced pelvic excursion (85 mm vs. 110 mm), increased torso pitch and a substantially lower number of rib fractures (1 fx, 0 fx, 4 fx) as intended. The predicted risk of rib fractures provided by the THOR dummy using the C_{max} injury criterion was lower than the actual injuries observed in the PMHS tests (especially in restraint condition A). However, the THOR dummy was capable to discriminate between the two restraint scenarios.

INTRODUCTION

Life expectancy has increased worldwide thanks to continuous improvements in medical technology, housing, nutrition and education (Kirkwood, 2008). Additionally, the number of births in several developed countries has declined in recent decades leading to an increase share of older people in the population (EuroStat, 2013): while 17% of Europeans were aged 65 and older in 2012, this proportion will raise to 28% in 2020 (European Commission, 2011). As people also tend to drive later in life, protecting older occupants is becoming a priority in many countries in the world. But developing restraint systems capable of preventing injuries to older occupants poses a significant challenge as elderly people are exposed to a higher risk of injury for a given magnitude of loading and the risk of a worse outcome for the same AIS-level injury is also higher (Kent et al., 2009). Recovering time is longer and usually the disability risk is higher compared to the younger population, regardless of body region (Schoell et al., 2016).

Identifying injury thresholds relies on biomechanical experiments in which different surrogates are used to represent the living human (animals, crash test dummies or ATD and Post Mortem Human Surrogates or PMHS). A recent review of the biomechanical experiments performed over the last decades showed that a substantial proportion of the studies have investigated the effects of advanced age on injury tolerances (Forman et al., 2015), largely focusing on the chest and ribcage. Despite all this research, it has been shown that the thorax continues to be the most critical body region for older car occupants as they present the highest share of AIS 3+ injuries (Wisch et al., 2017), which prompted the European Commission to fund the SENIORS (Safety Enhanced Innovations for Older Road userS) project aiming to improve the safe mobility of the elderly. SENIORS could be considered a follow-up of the research performed in the THORAX project, also funded by the European Commission, where the development and use of multi-chest deflection injury criteria with the THOR ATD in combination with the use of computational Human Body Models was explored (Davidsson et al., 2014) as an assessment tool to develop more effective restraint systems.

The need of incorporating factors that influence the injury occurrence (such as age or size of the individual) has been recognized in several studies that investigated how to use injury criteria with different types of ATD (Kent and Patrie, 2005; Laituri et al., 2005). With the recent finalization of the development and standardization of the THOR dummy, contemporary research has proposed chest injury criteria based on the multipoint chest deformation measuring capabilities of the ATD (Poplin et al., 2017), including age modifiers in the injury functions. This research noted the need of validation of the proposed injury criteria with an independent experimental dataset of matching THOR and PMHS tests. With the aim of developing a robust injury risk function that would not be sensitive to experimental conditions (Kent et al., 2003; Petitjean et al., 2003), the aforementioned study included a comprehensive sample of matching THOR and PMHS tests including different types of restraints, several seating positions and impact speeds. However just a few cases in the dataset used incorporated an airbag, which is mandatory in all nowadays vehicles in developed countries.

In addition, while most of the previous research has addressed AIS 3+ thoracic injuries, it is expected that in the short future automated features in passenger cars can reduce crash speeds and consequently the associated injury severity.

Thus, the goal of the study was to develop experimental chest loading conditions that would cause up to AIS 2 chest injuries in elderly occupants in moderate-speed frontal crashes. The new set of experimental data was also intended to be used in the benchmark of existing thoracic injury criteria in lower-speed collision conditions. To this end, six frontal sled tests with the THOR ATD and six elderly (>65 years old) PMHS were performed at 35 km/h. Two different restraint configurations were evaluated, which served to assess the robustness of proposed injury criteria.

METHODS

Test setup and conditions

The test fixture was designed to approximate the seating position of a front seat passenger car occupant using a simplified geometry (**Fehler! Verweisquelle konnte nicht gefunden werden.**). All components were designed and constructed so that any laboratory could reproduce the tests with the exact same setup. The design objective was twofold: to facilitate the use of the test fixture so that additional test data produced by any other laboratory could be added to the experimental data included here and to facilitate the development of Finite Element (FE) models of the experiments. To that end, the fixture consisted of a rigid inclined seat in the rear-forward direction (designed to match the sagittal displacement of the occupant pelvis with that obtained in a production seat, (Pipkorn et al., 2016)), a rigid footrest and an adjustable backrest made out of three segments of metal wire.

The motion of the occupant was arrested by a non-retractor, force-limited three-point seat belt. The force-limiting characteristic was provided by a generic load-limiter device that provided a nominal constant force due to the controlled deformation of a set of calibrated metal strips (Shaw et al., 2014) (**Fehler! Verweisquelle konnte nicht gefunden werden.**). The seat belt was adjusted before test and given a 50 N pretension that ensured a good fit on the occupant's chest removing any slack. The height and lateral position of the D-ring were adjusted to provide a similar set of conditions across the different occupants' anthropometries (height of the External Auditory Meatus of the occupant, 100 mm outward from the right acromion of the occupant). A pre-inflated (vented at t=0 ms) airbag attached to a rigid frame was also used to restrain the forward motion of the occupant. The airbag forward position was set so that the occupant's chest was in initial contact with the inflated bag. After all these adjustments, two sets of restraint conditions were implemented as indicated in *Table 1*. Condition A was used with PMHS1 – PMHS3 and Condition B was used with PMHS4-PMHS6. Under the corresponding restraint conditions, test subjects were exposed to a frontal impact following a trapezoidal deceleration with a plateau about 14 g that resulted in a 35 km/h delta-v. The THOR-M dummy was exposed to similar test conditions to have paired PMHS-THOR tests that could be used to benchmark injury criteria.

The geometry and components of the test fixture and the restraint conditions were proposed based on preliminary numerical studies and experimental tests with the THOR-M dummy targeting a low-severity impact condition that would not cause above AIS 2 level chest fractures (AAAM, 2008). The setup modification between condition 1 and condition 2 was based on a computational parametric study that used the THUMS human body model as occupant surrogate.

Test subjects

The THOR M dummy and six PMHS were exposed to matching impact conditions in the study. The THOR M dummy used in this study corresponds to the metric version of the THOR 50th percentile male dummy, including the SD-3 shoulder assembly (Parent et al., 2013). All subsesquent references to THOR in this manuscript pertain to this specific ATD model. As for the PMHS, six male elderly surrogates were chosen for this study. Computed tomography (CT) scans were taken prior to test, to ensure that there were not previous conditions that could compromise the results of the study, and post tests to assist in the injury assesement. PMHS were also subjected to a post-test detailed autopsy. The main characteristics of the test subjects are included in *Table 1*. Procurement, handling and testing of the PMHS was done under the approval of the Ethical Commission for Clinical Research of Aragon (CEICA), which is the official body responsible for assessing all research projects involving human subjects in the region of Aragon (Spain).

The overall seating procedure was similar for the ATD and PMHS and consisted of the following steps:

1) Position the occupant pelvis on the midline of the seat.
2) Ensure that the H-point or greater trochanter of the occupant was aligned with the H-point defined in the seat.

3) Adjust the position of the foot rest so that the femur angle was around 11 degrees and the tibia angle was around 50 degrees.
4) Adjust the position and tension of the seat back wires so that the occupant torso angle measured over the sternum was about 30 degrees.

Instrumentation

Upper shoulder and lower shoulder belt tension were measured at defined positions of the belt webbing. Lap belt tension was measured bilaterally. The THOR M dummy was equipped with the conventional instrumentation that included the IRTRACC sensors to measure the deformation of the ATD chest at four different locations. AIS3+ thoracic injury risk was estimated using the injury criterion based on the calculation of the maximum resultant deformation (C_{max}) given by the four IRTRACC and was adjusted to a 65 YO occupant (Crandall, 2013). In the PMHS tests, triaxial accelerometers were rigidly attached to the top of the head, two locations on the thoracic spine (upper and middle regions, nominally T1 and T8), lumbar spine (L2) and the pelvis. Triaxial Angular Rate Sensors (ARS) were added to the head, upper spine and pelvis. All sensor data were captured at 10,000 Hz with an external data acquisition system (PCI-6254, National Instruments; Austin, TX). Tests were recorded by a lateral and a frontal high-speed imager at 1,000 Hz.

Marker position. Data processing

Retro-reflective spherical markers were attached to selected locations on the head; upper, mid and lower sections of the spine; acromion bilaterally; pelvis; hip joint bilaterally and on other selected landmarks. Corresponding locations of the ATD were also tracked during the tests. Marker clusters were used on the head, spine and pelvis of the PMHS. Clusters were attached using rigid, low-weight aluminum plates to the corresponding anatomical structure. Other locations were tracked using single markers glued directly to the bandage covering the subject. Kinematic data were collected at 1,000 Hz using an optoelectric stereophotogrammetric system consisting of 10 cameras (Vicon, TS series, Oxford, UK). The system captured the position of the markers within a calibrated 3D volume. A calibration procedure, performed prior to testing, estimated the optical characteristics of each camera and established its position and orientation in a reference coordinate system. The trajectory of each marker was recorded and smoothed through a rigidity constraint using the least squares pose (LSP) estimator (Cappozzo et al., 2005; Chiari et al., 2005; Della Croce et al., 2005; Leardini et al., 2005). A global coordinate system (GCS) was defined at a laboratory fix location. A local coordinate system (LCS) moving with the test buck was defined with origin at the front right corner of the seat following SAE J211 indications (Society of Automotive Engineers, 1998). Local x axis pointed forward and it was coincident initially with the frontal anatomical axis of the occupant. The vertical z axis pointed upwards (opposite to ground) and the y axis was defined to form a right-hand oriented coordinate system. Unless otherwise indicated, displacement data are expressed with respect to this LCS. A photogrammetric algorithm within the Vicon Nexus software package (Nexus 1.8.5, Vicon, Oxford, UK) reconstructed the 3D position of each target for each video sample increment from the multiple 2D camera images.

RESULTS

First round of THOR and PMHS tests

Table 2 shows some of the most relevant results obtained in the THOR tests using the first set of restraint conditions. Maximum resultant chest deformation was measured at the Upper Left IRTRACC of the ATD (33.7±1.2 mm) followed by the Lower Left IRTRACC. This deformation resulted in a 26.6% probability of a 65-YO of sustaining AIS3+ injuries.

Figure 1 compares the time history of the shoulder seat belt force between the THOR and the PMHS. The plot shows that the interaction with the seat belt of the ATD and the PMHS was similar in terms of timing and magnitude, with the exception of the third PMHS, who was considerably lighter than the dummy. THOR response corridor lies in between the responses of PMHS 1 and PMHS 2. In addition, the peak lap belt force measured in the dummy tests was almost equal to the one measured in the aforementioned PMHS tests (3.8 kN (THOR) vs. 3.7 and 3.8 kN (PMHS)). The forward peak excursion of the THOR head center of gravity (CG) was considerably higher than those observed in the PMHS tests (464.8 mm vs. 283.2 mm, average of PMHS 1 and PMHS 2), but these differences in peak displacements decreased along the spine up to the lower extremities, where the displacement measured at the H-point was similar between the THOR and PMHS 1 and PMHS 2. The stature and weight of PMHS 3 resulted in very different magnitudes of seat belt forces and displacements. The autopsy revealed that the three PMHS sustained more than 10 rib fractures (resulting in flail chest in two occasions, which is ranked as an AIS 5 injury) (see Figure 2) as result of the deceleration despite that the THOR prediction of AIS 3+ chest injuries was 26.6%.

Parametric study with the THUMS model

A parametric study using computer simulation was carried out using the THUMS model. The objective of the study was to identify which features of the test setup and procedure could be improved to decrease the number of rib fractures. The number of fractured ribs was estimated by comparing the predicted ultimate strain with experimental data from rib cortical bone tests, adjusted by age (Forman et al., 2012). When the THUMS model was positioned and adjusted to mimic the tests of PMHS 1 and 2, and after showing a good correlation with the test measured seat belt forces, seat forces and displacements, predicted only a 6% risk of sustaining AIS 3+ chest injuries. THUMS predicted that the maximum X deflection occurred at the lower right aspect of the ribcage, where the THOR ATD measured the lowest resultant chest deflection (39.5 mm vs. 10.3 mm). Despite these differences in chest deformation, the model was considered to provide a reasonable approximation to the physical tests.

The variables included in the parametric study were: the forward position of the D-ring, forward position of the seat buckle, force-limit magnitude, airbag pressure, airbag venting trigger time, airbag height with respect to the occupant and seat friction. The combination of lowering the force limit and moving rearward the D-ring resulted in a 30% predicted reduction of having 2+ fractured ribs. The effect of modifying the airbag pressure or the venting trigger time was unclear, although contributed slightly to reduce the predicted injury risk. Increasing the friction of the seat surface alone reduced the forward displacement of the pelvis by 14%. All these factors together resulted in a 0% prediction of 2+ fractured ribs. Consequently, this setup was chosen to be used in the second series of THOR and PMHS tests. Compared to the initial set of testing conditions, the final setup modified the following parameters:

- Seat belt system: lowered the force limit to 2 kN; moved rearward 100 mm the position of the D-ring.
- Airbag system: lowered the height of the airbag by 25 mm; filled the airbag at 11 kPa.
- Seat: increased the friction of the surface of the seat.

In addition to these changes in the test setup, it was also observed that the seating procedure followed in the initial round of tests and the different geometries of the THOR ATD and the PMHS (or, correspondingly, the THUMS model), allowed for uncontrolled variability in the initial position between the surrogates (**Fehler! Verweisquelle konnte nicht gefunden werden.**,) including also inter-PMHS variability. Therefore, a more detailed seating procedure was designed in which additional control points for the initial positioning of the THOR dummy and the PMHS were considered. This seating procedure facilitated that some instances of the test fixture could be adapted to the anthropometries of the specific PMHS exposed to the test. In particular, the forward position of the PMHS head (X_{EAM_PMHS}) and height of the airbag ($Z_{Steering\_wheel\_PMHS}$) coordinates were modified according to the following formulae (final values are provided in *Table 1*):

$$\frac{X_{EAM\_THOR} - X_{H-point\_THOR}}{Sitting\ height_{THOR}} = \frac{X_{EAM\_PMHS} - X_{g\_troch\_PMHS}}{Sitting\ height_{PMHS}}$$

$$\frac{Z_{Top\_THOR} - Z_{Steering\_wheel\_THOR}}{Sitting\ height_{THOR}} = \frac{Z_{Top\_PMHS} - Z_{Steering\_wheel\_PMHS}}{Sitting\ height_{PMHS}}$$

Last, a major difference between the first and second series of tests (applicable only to PMHS) was to provide a more slouched initial position in the second series. This position was achieved by setting the angle measure between the spinous process of T1 and T12 to be around 10 degrees and was considered more reliable than measuring the sternum angle. Unfortunately, this angle was not measured in the initial test series.

Second round of THOR and PMHS test

As result of the above-mentioned changes, the C_{max} value was reduced to 27.6 that corresponded to an estimated 10.7% AIS 3+ chest injury risk. The deflection of the THOR thorax was spread over the two left and the upper right IRTRACC sensors, while the lower right one provided the minimum resultant deflection as in the previous test round. Lowering the load limit of the seat belt and the airbag pressure did not influence the peak forward displacement of the CG of the ATD head (469.3±21.5 mm) compared to the one obtained in the first round (464.8±10.8 mm). Similar observations were made for the upper spine and H-point locations. The lap belt seat force increased by 400 N, despite of the greater friction coefficient of the seat surface. The time history plot of the shoulder belt force shown in Figure 1 illustrates that the interaction between the shoulder belt and the two different surrogates was even more similar in terms of the phasing and magnitude than in the first restraint condition. This can be likely attributed to a more detailed positioning of the subject that included the scaling relationships shown above.

These changes affected substantially the injury outcome of the three PMHS test. In this case, the reduction in airbag pressure and load limit caused an increased head forward excursion (average excursion: 495.9 mm) that was greater than the one observed in the THOR dummy despite the PMHS were shorter subjects. The displacement of T1 followed the same trend, but the H-point forward excursion remained within the values observed using the initial restraint conditions (average excursion: 83.4 mm) likely indicating that the increased friction of the seat surface was more effective than in the case of the THOR dummy. The additional restraint provided by the seat surface caused the lap seat belt forces to be around 600 N (in average) lower than those observed in the first round of PMHS tests, contrary to what had happened in the ATD. As for the rotation of the head, the new set of restraint conditions resulted in an increased flexion rotational speed of the head compared to the initial tests. The major difference however was found in the injury outcome of the three subjects as PMHS 4 sustained only one rib fracture, no injury was found in the case of PMHS 5 and PMHS 6 received four rib fractures. All these results are included in Table 2.

DISCUSSION

Development of chest injury risk functions for the THOR M ATD

One of the primary goals of the EU funded SENIORS project was to develop injury risk functions and testing methods that could be used to improve the protection of elderly road users. Given the high incidence of thoracic injuries in the field (Scarboro, 2014) and despite the large number of biomechanical studies performed with elderly subjects, assessing the chest injury risk estimation provided by the THOR dummy was considered the natural start. Contemporary research has proposed several thoracic injury criteria to be used with THOR, taking advantage of the multi-point deflection characteristics of this ATD (Davidsson et al., 2014; Hynd et al., 2013; Poplin et al., 2017). In an effort to develop a robust injury risk function, the last mentioned study combined diverse test conditions to overcome the difficulties associated to initial positioning, restraint type and impact conditions (Kent et al., 2003; Petitjean et al., 2003). Poplin and coauthors proposed to use the so-called C_{max} and PC Score as the most appropriate injury metrics for the THOR ATD, given that both were qualitatively equivalent. The present study is completely independent from the tests used in the development of the proposed injury risk functions and therefore can serve to benchmark them. Only the C_{max} metric adjusted by age has been used here. Two of the PMHS tests resulted in flail chest (PMHS 1 and PMHS 2), which corresponds to an AIS 5 injury while the three PMHS in the second series received only AIS 1, no injuries and AIS 3 injuries. The C_{max} values obtained in the THOR tests were 33.7 and 27.6 (26.6% and 10.7% injury risk respectively), which seem to underestimate and overestimate the results observed in the PMHS tests. However, the injury metric was capable to capture the injury severity reduction later observed in the PMHS tests. Thus, the THOR ATD and C_{max} were sensible to these restraint changes despite existing differences in the prediction of the kinematics (displacements and angular rate of the head) as seen in Table 2.

Test surrogates

Both the physical THOR dummy and the FE THUMS human body model failed to capture correctly the exact kinematics, dynamics and injury outcome observed in the PMHS tests. These differences can be, at least, partially explained by the difficulties encountered in defining a matching initial position between the three different types of surrogates provided that the anthropometry and landmarks were so different between them. This is particularly true between the ATD and the human surrogates. The challenge of defining a common initial position exists also within the PMHS group given the differences in anthropometry across subjects. Ideally a method that would allow documenting the initial position of internal bony landmarks and scaling the restraint geometry accordingly would be the preferred option. However, this method would require knowing the seating posture of the PMHS in advance or measuring it during the preparation of the test, which may constitute a cumbersome task to be added to the usually long procedures associated to PMHS testing.

However, it is remarkable that both the THOR dummy and the THUMS model were capable of capturing the differences between restraint conditions 1 and 2 and predicted injury severity reductions that were later corroborated by the PMHS tests.

Influence of seating procedures and test setup

Note that, in the case of the THOR tests, the initial position of the dummy was not essentially different between the two test rounds and therefore the kinematic and dynamic changes observed can be attributed only to the setup change. This finding indicates that the injury reduction outcome predicted by the human body model in the parametric study (in which the initial position was fixed across the different restraint conditions) was supported also by the physical tests run with the THOR M dummy.

As for the PMHS tests, the substantial injury reduction observed in the second round of tests is likely due to the combination of the changes in the restraint conditions and an updated initial positioning protocol that resulted in an increased torso angle with respect to the vertical direction. This torso angle allowed for a greater excursion of the head and for, in general, more favorable overall kinematics of the occupant (Adomeit and Heger, 1975; Kent et al., 2011; Lopez-Valdes et al., 2014). The examination of the high-speed video captures showed that in the initial set of conditions, the trunk and head of the PMHS moved forward almost as a rigid body while in the second case the pelvis is restrained by the seat belt and the additional friction of the seat causing the torso to pitch forward in much more favorable restraint interaction with the thorax structures.

A recent study exposing small elderly female PMHS to what was intended to be a minimally injurious frontal crash test (in conditions similar to the ones used in this paper, without the airbag) also resulted in an unexpectedly high number of rib fractures (Shaw et al., 2017). Unfortunately, the measurement of 3D chest deflection could not be done in our tests due to the pre-inflated airbag and no further comparison can be established with Shaw’s study.

Limitations of the study

Some of the instrumentation did not work correctly during the tests. In test 1743 (THOR), the motion capture system failed during the tests and the displacement results shown here include only the other two tests. The pressure sensor of the airbag malfunctioned in test 1761 and the initial pressure could not be measured.

While the positioning of the PMHS in the first round of tests was performed according to normal procedures in these type of tests, the differences observed between the outcome in the THOR and PMHS tests caused the positioning procedure to be reviewed to obtain more similar initial positions between the ATD and the humans. The new protocol included a more quantitative approach to positioning finding corresponding landmarks and scaling relationships between the surrogates. Unfortunately, several of the positioning parameters used in round two were not registered in round one as they were not considered necessary for the tests.

PMHS 3 is clearly an outlier in this study in terms of age and anthropometry, and so are the results observed from this test. While it would not be correct to establish a direct comparison between this PMHS and the THOR dummy, we have considered that it was worth reporting the results observed for this subject as, if corrected by age, its anthropometry is not far from the predicted anthropometry of a 94 year old (average height: 159 cm; average weight: 59.7 kg) (Perissinotto et al., 2002). Interestingly, this subject sustained 15 rib fractures and most of them occurred in the posterior aspect of the rib cage, bilaterally and close to the costovertebral junction. This chest injury pattern is uncommon and was totally different from what was observed in the other two PMHS tests of this series. This particular test subject presented a very stiffened spine due to the formation of osteophytes that produced a prominent kyphosis of the thoracic spine. Whether this characteristic is related to the exhibited injury pattern is unknown.

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Table 1 Set up conditions, initial position and anthropometry of test subjects. XYZ coordinates are given with respect to a coordinate system placed at the intersection of the fore/after midline of the seat and the line joining the bilateral defined position of the H-point seat.

	THOR	PMHS1	PMHS2	PMHS3	THOR	PMHS4	PMHS5	PMHS6
	Condition A				Condition B			
Test	1743, 1744, 1745	1761	1763	1765	1961, 1962, 1968	1969	1970	1971
Seat belt pretension (kN)	0.05	0.05	0.05	0.05	0.05	0.05	0.05	0.05
Force limit (kN)	2.5	2.5	2.5	2.5	2.0	2.0	2.0	2.0
Airbag pressure (kPa)	14	--	14	13	11	11	11	11
D-ring X position (mm)	-320	-320	-320	-320	-420	-420	-420	-420
D-ring Y position (mm)	260	§	§	§	303	317	290	276
D-ring Z position (mm)	-723	§	§	§	-740	-692	-726	-668
Airbag X position *	371	**	**	**	371	411	422	434
Airbag Y position *	1	1	1	1	6	6	6	6
Airbag Z position *	-707	-707	-707	-707	-682	-637	-642	-592
Seat friction	No	No	No	No	Yes	Yes	Yes	Yes
	Initial position							
H-point R/L (mm)	-1/2	-5/-11	0/0	-10/-5	-3/3	4/4	3/4	-4/-4
Sternum angle (deg)	33.0	22.5	26.0	24.0	38.2	31.0	40	27.0
T1/T12 angle (deg)	--	--	--	--	--	10.6	12	10.0
Shoulder belt angle (deg)	25	28	19	22	20.6	18.0	20.6	23.5
	PMHS characteristics and anthropometry							
Age	--	74	68	94	--	74	63	73
Sex	Male	Male	Male	Male	Male	Male	Male	Male
Height (cm)	175	167	184	156	175	170	174	167
Sitting height (cm)		74.5	79.0	72.0		76.9	82.4	78.5
Weight (kg)	82	66.0	76.0	34.0	82	74	67	62
Cause of death	--	Hepatic infection	Liver infection	Prostatic cancer	--	Liver cancer	Ling cancer	Cardiopathy

§ Position of the seat belt D-ring was personalized to each occupant but was not documented during the test.
* XYZ coordinates of the marker on top of the steering wheel assembly ** The airbag structure was positioned as close to the occupant as possible but ensuring no initial contact, position not documented during the tests.

Table 2 Selected results from test series, including THOR and PMHS tests.

	THOR	PMHS1	PMHS2	PMHS3	THOR	PMHS4	PMHS5	PMHS6
	Condition A				Condition B			
Test	1743, 1744, 1745	1761	1763	1765	1961, 1962, 1968	1969	1970	1971
Shoulder seat belt peak force (kN)	2.6±0.1	2.3	2.7	1.8	2.1±0.05	2.0	1.8	1.9
Lap seat belt peak force (kN)	3.8±0.1	3.7	3.8	1.7	4.2±0.3	3.3	2.8	2.2
Head CG X displacement (mm)	464.8±10.8	296.7	269.7	194.7	469.3±21.5	530.9	506.7	450.1
T1 X displacement (mm)	322.5±11.1	213.2	202.4	141.4	325.0±23.5	422.2	359.2	375.5
H-point X displacement (mm)	111.6±2.8	95.4	104.8	36.6	110.4±2.7	87.5	70.8	91.8
Head max ARS X (deg/s)	-178.8±37	658.4	293.7	281.6	-120.2±31	534.9	376.8	158.9
Head max ARS Y (deg/s)	-1307.5±23	-1346.8	-1238.8	-737.1	-1131.0±2	-1759.8	-1533.3	-1424.9
Head max ARS Z (deg/s)	325.2±64	255.8	593.4	171.7	411.3±99	350.9	495.7	472.4
UL Max resultant	33.7±1.2	--	--	--	27.6±0.6	--	--	--
UR Max resultant	18.9±0.9	--	--	--	26.7±0.4	--	--	--
LL Max resultant	28.3±1.8	--	--	--	27.1±0.7	--	--	--
LR Max resultant	10.3±0.2	--	--	--	9.6±1.0	--	--	--
C_{max}	33.7	--	--	--	27.6	--	--	--
p(AIS3+)$_{Cmax}$ 65 YO (%)	26.6	--	--	--	10.7	--	--	--
Rib fx	--	10	14	15	--	1	0	4
Sternum fx	--	Yes	Yes	No	--	Yes	No	No
Other injuries	--	--	C7-T1 interspinous ligament tear	--	--	--	--	--

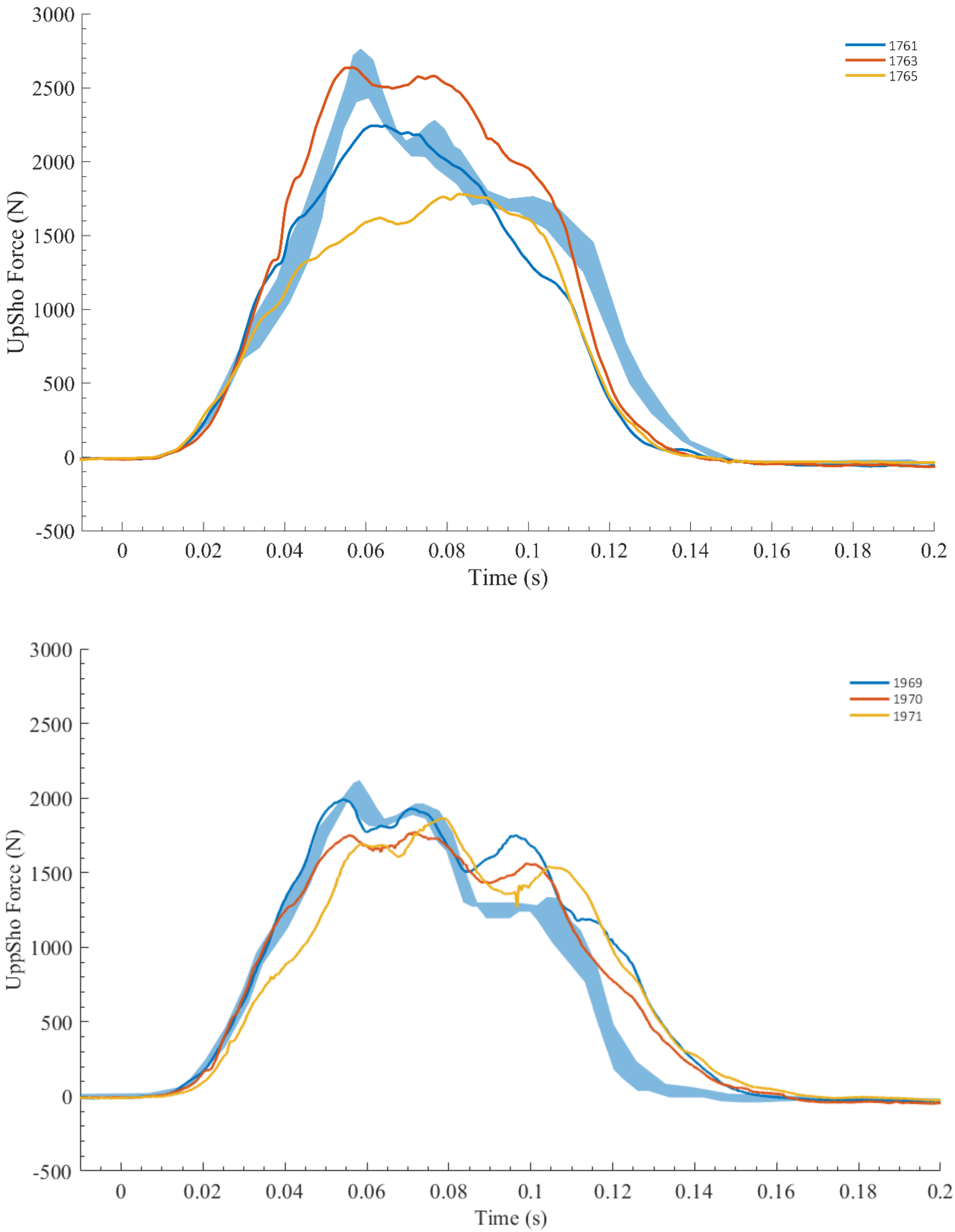

Figure 1 Time history of the upper shoulder seat belt force. Top: first series of PMHS tests (solid lines) and corridor response of corresponding THOR tests (shaded blue area). Bottom: second series of PMHS tests (solid lines) and corridor response of corresponding THOR tests (shaded blue area).

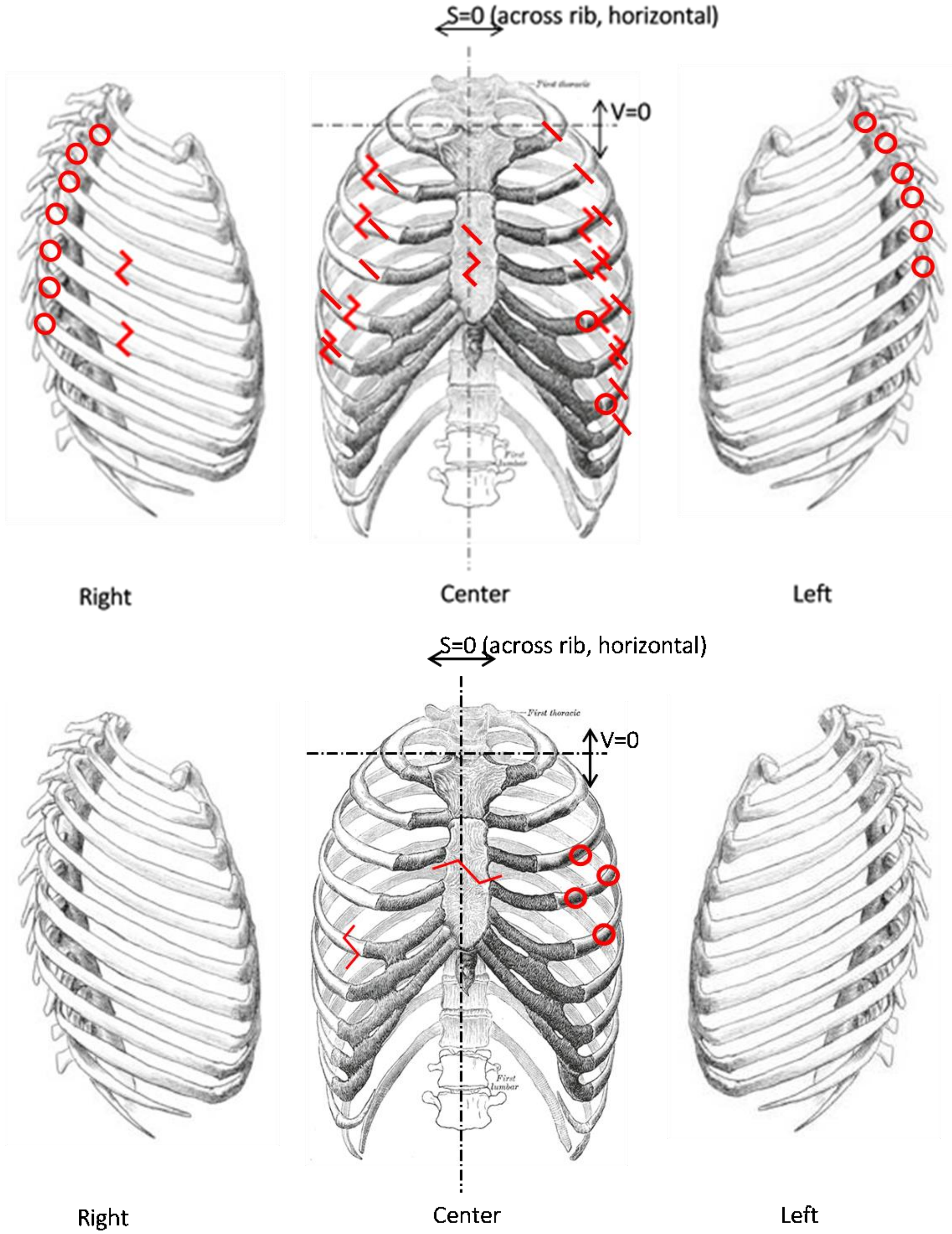

Figure 2 Approximate location of the rib and sternal fractures observed in the post-test PMHS examination. First PMHS series above (PMHS1: angled line; PMHS2: straight line; PMHS3: circle). Second PMHS series below (PMHS4: angled line; PMHS5: straight line; PMHS6: circle