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In-vivo analysis of thoracic mechanical response under belt loading: The role of body mass index in thorax stiffness

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ABSTRACT

Thoracic injuries are a major cause of mortality in frontal collisions, especially for elderly and obese people. Car occupant individual characteristics like BMI are known to influence human vulnerability in crashes. In the present study, thoracic mechanical response of volunteers quantified by optical method was linked to individual characteristics. 13 relaxed volunteers of different anthropometries, genders and age were submitted to non-injurious sled tests (4 g, 8 km/h) with a sled buck representing the environment of a front passenger restrained by a 3-point belt. A resulting shoulder belt force was computed using the external and internal shoulder belt loads and considering shoulder belt geometry. The mid sternal deflection was calculated as the distance variation between markers placed at mid-sternum and the 7th vertebra spinous process of the subject. Force-deflection curves were constructed using resulting shoulder belt force and midsternal deflection. Average maximum chest compression was $7.9 \pm 2.3\%$ and no significant difference was observed between overweight subjects (BMI ≥ 25 kg/m^2) and normal subject (BMI < 25 kg/m^2). The overweight subjects exhibited significantly greater resultant belt forces than normal subjects (715 ± 132 N vs. 527 ± 111 N, $p < 0.05$), higher effective stiffness (30.9 ± 10.6 N/mm vs. 19.6 ± 8.9 N/mm, $p < 0.05$) and lower dynamic stiffness (42.7 ± 8.71 N/mm vs. 61.7 ± 15.5 N/mm, $p < 0.05$).

1. Introduction

Although significant improvements have been achieved in mitigating road traffic fatalities, frontal impacts play a predominant role in the frequency of road traffic fatalities as they account for up to 46% of the mortality (Klanner, 2001). Statistics show that in France thoracic injuries were the first cause of death among car occupants in frontal collisions and the main source of serious casualties of elderly occupants (Lafont and Laumon, 2003; Ndiaye and Chiron, 2009). Moreover, if we focus on moderate injuries sustained by the elderly population, rib and sternum fractures are mainly observed (Ndiaye and Chiron, 2009).

First studies on Post Mortem Human Surrogates (PMHS) focused on generating thoracic force-deflection corridors using a hub-impact test condition (Kroell and Schneider, 1971; Lobdell et al., 1973; Nahum et al., 1975; Lau and Viano, 1986). These corridors were used in the development of frontal impact dummies. A statistical relationship was established between midsternal compression and thorax injury risk (Kroell and Schneider, 1971; Mertz et al., 1991). However, force-deflection corridors of the thorax were found to be strongly dependent on the load distribution (Kent et al., 2004), and particularly on the anatomical area that supports the load (Shaw et al., 2007). Moreover, thoracic mechanical response depends strongly on the geometrical characteristics of the ribcage and on its biological material properties (Kent et al., 2005; Gayzik et al., 2008). Authors agree that the deformation of the rib cage under an antero-posterior loading is divided into rib rotation relative to the vertebrae and sternum, and rib deflection (Kent et al., 2005). More accurately, it is assumed that according to the rib orientation, the applied force causes rib deformations and rotations at various degrees (Kent et al., 2005). Specifically, for an individual with a “vertical” rib cage, an applied antero-posterior load produces rotations of the rib joints and bone deformation whereas, for individual with a “horizontal” rib cage, the load force acts straight in the plane of the ribs and implies primarily bone deformation. A link was found between the initial rib slope and the amount of rotation and deformation (Vezin and Berthet, 2009). Thus, it seems that geometrical characteristics of the ribcage may predispose ribs to fracturing.

In addition, it was well established that tolerance of the human thorax under dynamic loading decreases as age increases (Ndiaye and Chiron, 2009; Kroell and Schneider, 1971; Zhou et al., 1996; Kent et al., 2005). Physiological phenomena of ageing could

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be described as the combination of mechanical and structural modifications associated with a decrease in chest deflection tolerance. The elastic modulus and the tensile strength of human bones decrease with age (Zhou et al., 1996). Similarly, a decrease in the tensile strength of hyaline rib cartilage and cortical thickness in the rib sections was observed (Takahashi and Frost, 1966; Yamada, 1970; Kemper et al., 2007). The age-related changes in thorax shape were also studied and authors assumed that the ribs became more horizontal with age and the slope of ribs in sagittal plane decreased with age (Oskvig, 1999; Kent et al., 2005; Gayzik et al., 2008). Yet, some authors also found a correlation between BMI and the thorax shape (Berthet et al., 2005; Gayzik et al., 2008). Body Mass Index (BMI) was associated with increased risk of mortality and increased risk of severe injury with a predominant occurrence for rib fractures and pulmonary contusions (Boulanger et al., 1992; Mock et al., 2002). Obese subjects were linked to unfavourable kinematics (greater head and pelvis excursions) and may be linked to higher chest compression (Forman et al., 2009). In the vast majority of developed countries, the prevalence of overweight and obesity in older adult population is very high (Gutierrez-Fisac et al., 2004).

Thus, in order to study the variability of the population in age, anthropometry and for both genders, in vivo experiments seem necessary. Except early in-vivo experimental studies (Armstrong et al., 1968; Lobdell et al., 1973; Stalnaker et al., 1973) which brought knowledge on the human body tolerance, most of them were used to analyze the physiology influence on the body response to dynamic loading—muscle tone (Patrick, 1981; Backaitis and St-Laurent, 1986; Kemper et al., 2011)—or to assess the response of specific populations such as children for non-injurious levels (Sandoz et al., 2009; Arbogast et al., 2009a, 2009b). In the present paper, we present the analysis of thoracic mechanical response under belt loading of adults of various anthropometries and age subjected to a low deceleration pulse (Poulard et al., 2011). In particular, the influence of overweight ($BMI \geq 25$ kg/m^2) on thoracic mechanical response will be assessed.

2. Materials and methods

This in vivo protocol was reviewed and approved by the French ethical committee Comité de Protection des Personnes Sud-Est II at Lyon in February 2011.

Fig. 1. Sled configuration based upon a standard sedan car environment of a front passenger restrained by a 3-point belt.

2.1. Test device

Ifsttar shock sled was set up with a standard sedan car environment of a front passenger restrained by a 3-point belt (Fig. 1). Posterior support consisted of a belt band tensed between the two backseat rods. Weight was added to the sled according to the weight of the subject (up to 110 kg) in order to maintain a constant sled mass (640 kg). The propelling system of the sled was chosen to allow low impact velocities with a good repeatability in pulses and to limit the maximum impact speed; it was made of two rubber bands of diameter 20 mm mounted on each side of the sled.

2.2. Deceleration pulse

As described in Poulard et al. (2011), the deceleration pulse was chosen between 2 pulses measured on 2 public demonstrators and it was found similar to the one of a recent study (Arbogast et al., 2009a). The sled was pulled rearward by an electrical winch until reaching 165% of the two rubber bands' initial length so that an impact speed of 8 km/h was reached. The deceleration system was obtained by laminating a polyurethane tube by a spear ended in an olive shape (hardness 95 SHA, 46.5 mm diameter spear). The sled was stopped with a maximum deceleration of 4 g during 120 ms.

2.3. Tested subjects

13 volunteers were tested. They were between 19 and 65 years of age and consisted of both males and females. They were chosen on criteria of height based upon anthropometry charts (Jürgens et al., 1990) and various BMI. Volunteer characteristics are listed in Table 1.

For physiological issues, subjects with existing endocrine disorders, mediastinal pathologies, previous or current diseases of the head, neck, spine were excluded from the tests. Only volunteers over 40 years of age with no osteopenia or osteoporosis (total femoral and lumbar $Tscore > -1$) were included in sled testing experiments. Prior to the testing, a medical doctor conducted an examination of each subject to confirm eligibility.

Two sled tests were performed on each volunteer. For each test, they were asked to adopt a relaxed posture.

2.4. Instrumentation

The longitudinal deceleration of the sled was recorded by a sled-mounted uniaxial accelerometer. Three belt webbing load cells were installed on the external end (FB3) and on the internal end (FB4) of the shoulder belt and on the external side of the lap belt. Maximum values of FB3 and FB4 were noted $FB3_{max}$ and $FB4_{max}$ respectively. Reaction forces were measured by four uniaxial load cells

Table 1
Anthropometric characteristics of volunteers.

ID	Gender	Age (yo)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Midsternal thickness (mm)
Normal subjects						
200	F	22	176	58	19	179
607	M	32	164	57	21	223
862	F	41	161	58	22	224
806	M	21	196	88	23	217
869	M	19	167	64	23	226
733	M	34	198	92	23	233
957	F	23	155	58	24	187
225	M	32	185	83	24	210
Average	–	*28*	*170*	*70*	*22*	*212*
SD	–	*8*	*16*	*15*	*2*	*19*
Overweight subjects						
329	M	38	173	74	25	213
948	M	25	169	73	26	225
118	M	65	167	74	27	234
442	M	35	176	86	28	258
370	F	20	173	90	30	219
Average	–	*37*	*172*	*79*	*27*	*230*
SD	–	*20*	*4*	*9*	*2*	*17*
All subjects						
Average	–	*31*	*174*	*73*	*24*	*219*
SD	–	*13*	*13*	*13*	*3*	*20*
Min	–	*19*	*155*	*57*	*19*	*179*
Max	–	*65*	*198*	*92*	*30*	*258*

placed under each footrest and under the seat pan. Data were acquired at 20 kHz during 5 s and filtered using CFC Butterworth filters according to the standard norm SAE J211.

2.5. Stereovision

Markers were placed on volunteers at anatomical landmarks:

- Spine: spinous processes of C7, T1, T7, and L4 and all along the spine,
- Torso: on left and right acromion and clavicles, jugular notch (JN), mid-sternum (MS), xiphoid process (XP) and all along the sternum. Additional landmarks were additionally placed 5 cm and 10 cm to the left and right side of chest and on the inferior contour of the ribcage.

In addition, markers were placed on the sled in order to follow sled displacement.

Four high speed cameras were positioned around the sled for recording at 1000 fps the position of markers placed on subjects and sled (Fig. 2). Time zero was given by a flash visible by all cameras and synchronised with the data acquisition system. After the tests, the scene was calibrated by using a calibration object and the three dimensional coordinates of the subject markers were computed using the Direct Linear Transformation method also called stereovision (Abdel-Aziz and Karara, 1971). Accuracy of the optical method was assessed in a previous study (Poulard et al., 2011). 3D trajectories were filtered at CFC-60 Butterworth.

Anatomical landmarks were obstructed from view during some frames during the test, and thus, the 3D reconstruction by stereovision was not possible during these frames. Therefore the missing part of the trajectory was interpolated using other landmarks available at the same frame (see Appendices for more details).

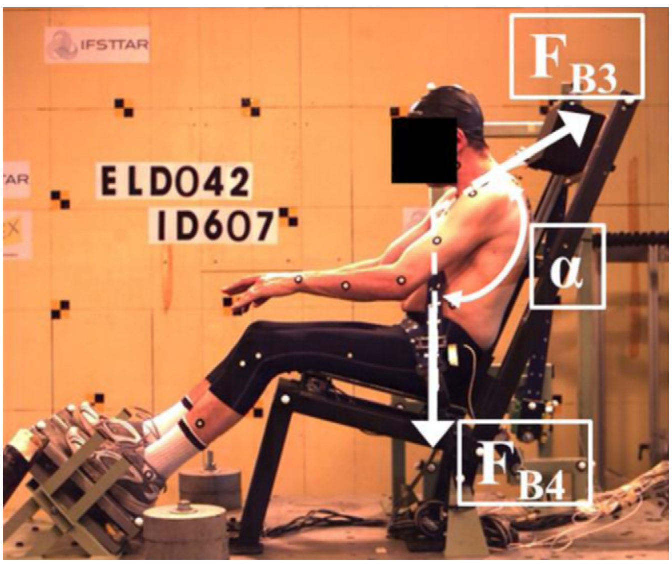

Fig. 3. Computation of the resulting shoulder belt load FRes as defined by (Eickhoff et al., 2011).

2.6. Data analysis

The mid sternal deflection (D) was calculated as the distance variation between markers placed at mid-sternum and at 7th vertebra spinous process of

Table 2
Thoracic mechanical characteristics obtained on volunteers.

ID	$FB3_{max}$ (N)	$FB4_{max}$ (N)	$FRes_{max}$ (N)	α_{ini} (°)	D_{max} (mm)	C_{max}	ES (N/mm)	DS (N/mm)
Normal subjects								
200	560	669	357	156	12.5	0.07	13.3	89.4
607	616	682	417	155	22.4	0.10	16.6	62.3
862	803	860	609	154	12.1	0.05	28.8	59.1
806	986	1113	526	148	14.1	0.07	30.9	46.2
869	1036	911	547	153	16.5	0.07	16.0	54.4
733	1288	1439	558	156	12.9	0.06	7.3	73.1
957	1090	1126	721	151	22.5	0.12	14.3	40.9
225	883	1023	544	149	10.7	0.05	30.0	68.9
Average	*908*	*978*	*535*	*153*	*15.5*	*0.07*	*19.7*	*61.8*
SD	*244*	*255*	*111*	*3.5*	*4.6*	*0.02*	*9.0*	*15.5*
Overweight subjects								
329	1045	1121	602	154	11.7	0.05	37.7	42.9
948	1092	1138	821	157	23.3	0.10	25.0	42.8
118	1067	1253	726	144	16.7	0.07	36.9	47.2
442	856	930	582	152	26.5	0.10	15.8	31.4
370	1181	1588	884	160	19.5	0.09	33.9	54.3
Average	*1048*	*1206*	*723*	*153*	*19.5*	*0.08*	*29.9*	*43.7*
SD	*138*	*275*	*131*	*6.9*	*4.3*	*0.01*	*9.5*	*9.6*
All subjects								
Average	*962*	*1066*	*607*	*153*	*17.0*	*0.08*	*23.6*	*54.8*
SD	*211*	*267*	*149*	*4.2*	*5.3*	*0.02*	*10.1*	*15.7*
Min	*560*	*669*	*357*	*144*	*10.7*	*0.05*	*7.3*	*31.4*
Max	*1288*	*1588*	*884*	*160*	*26.5*	*0.12*	*37.7*	*89.4*

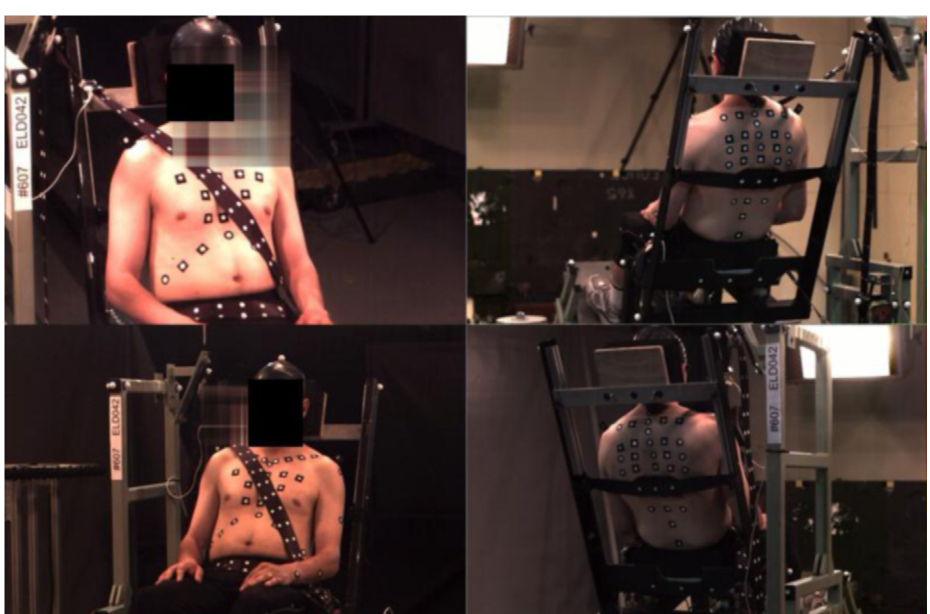

Fig. 2. Overview of camera location and field of view.

the subject. The initial zero of deflection was defined as the midsternal thickness measured after reaching 5% of the maximum value of external shoulder belt load (see midsternal thickness in Table 1). Midsternal compression C was defined as the ratio of the midsternal deflection to the midsternal thickness. Maximum values of D and C were noted D_{max} and C_{max} respectively.

Resulting shoulder belt force (FRes) is defined as $\mathrm{FRes}=\sqrt{\mathrm{FB3}^2+\mathrm{FB4}^2-2\times\mathrm{FB3}\times\mathrm{FB4}\times\cos(\pi-\alpha)}$and Fig. 3 as a combination of the external shoulder belt load (FB3), the internal shoulder belt load (FB4) and the angle made by the belt (α) calculated using the extreme markers of the belt (Eickhoff et al., 2011). A time-interpolation of FB3 and FB4 was performed for combining angles values and loads values. Maximum value of FRes was noted FRes_{max}.

FRes is a 2D approximation of the 3D resulting belt force acting on the thorax. Nevertheless, the obtained FRes force vector is correctly aligned with the midsternal deflection (D) vector (see Appendices for more details).

Force-deflection curves were built using FRes and midsternal deflection D. Dynamic Stiffness (DS) was computed as the interpolated ratio of the resulting shoulder belt load force FRes over the midsternal deflection D before $\mathrm{FRes}_{max}$was reached. Effective Stiffness (ES) was computed from the ratio of FRes over D when D_{max} was reached (Kent et al., 2005; Sandoz et al., 2009). While DS represents the thorax behaviour during the beginning of the load, ES represents its behaviour without taking account of this part.

The statistical analysis was carried out with R software for Windows ver. 2.14.2. Numerical variables were given as average ± standard deviation (SD). Groups were compared with permuted Student's t-test. Comparisons were considered as significant for $p<0.05$.

3. Results

3.1. Definition of an individual thoracic mechanical response

Two sled tests were performed for each volunteer. The force-deflection responses obtained in these two tests were found significantly different. The results of the test for which the highest Dmax was measured, was only considered. Thoracic mechanical characteristics are synthetized in Table 2. The initial shoulder belt angle α is noted αini.

3.2. Seat belt loads

Separate curves from overweight (BMI ≥ 25 kg/m^2, $n=5$) and normal subjects (BMI < 25 kg/m^2, $n=8$) for the external shoulder belt load FB3 and the resulting shoulder load FRes are shown in Fig. 4. At the beginning of the shock, 5% of FB3max was reached at 55 ± 5 ms ($n=13$). At an average time of 118 ± 8 ms ($n=13$), FB3 and FRes reached their peak. No significant difference ($p=0.18$) were found in $\mathrm{FB3}_{max}$ between overweight subjects and normal subjects. Average FRes_{max} was 723 ± 131 N ($n=5$) for overweight

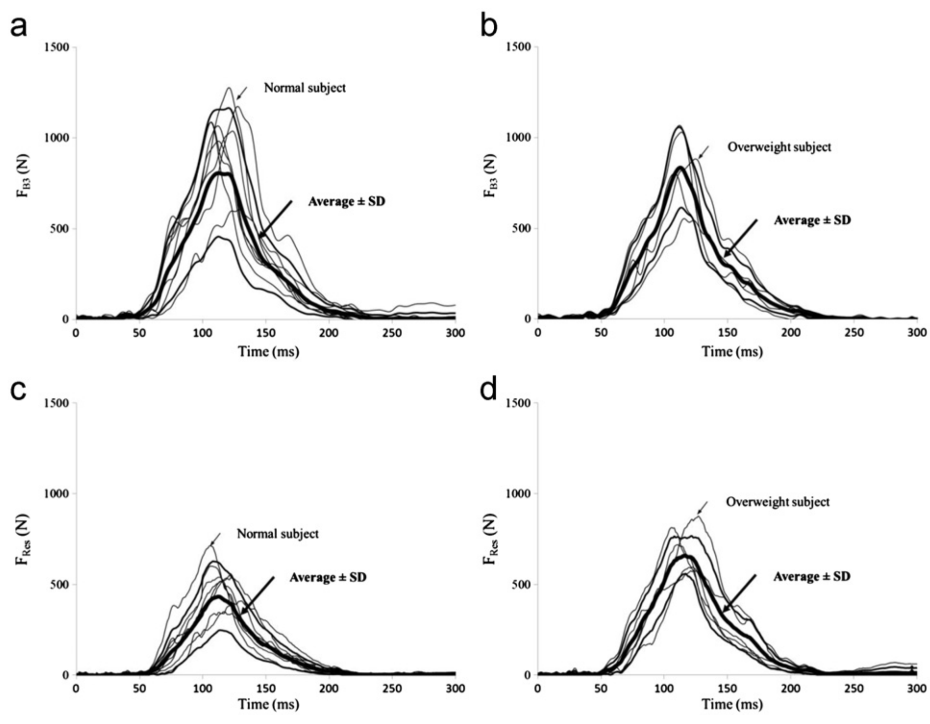

Fig. 4. Time history of the seat belt loads, external shoulder belt force FB3 and the resultant shoulder belt force FRes for normal group (a and c) and overweight group (b and d).

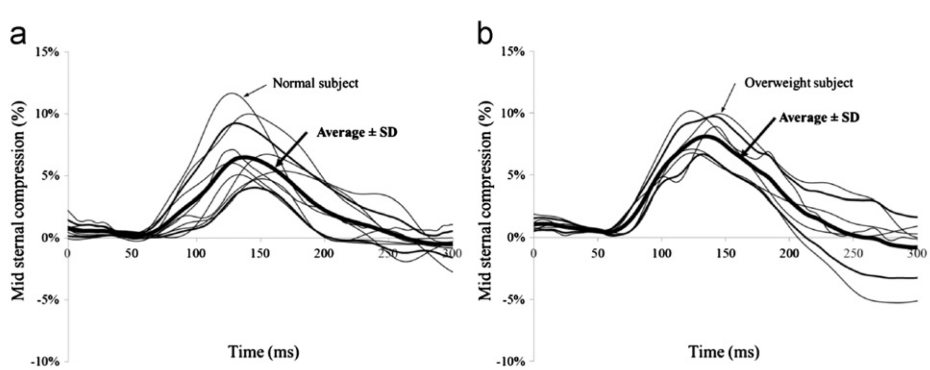

Fig. 5. Time history of midsternal compressions for normal group (a) and overweightgroup (b).

subjects, significantly higher ($p<0.02$) than 535 ± 111 N ($n=8$) for normal subjects.

3.3. Midsternal compressions

Time histories of mid-sternal compressions for the overweight group (a) and the normal group (b) are shown in Fig. 5. At an average time of 136 ± 15 ms, the midsternal compression reached its peak, C_{max}. There are no significant differences between overweight subjects (BMI ≥ 25 kg/m^2, $n=5$) and the subjects in the normal group (BMI <25 kg/m^2, $n=8$) in C_{max} ($p=0.41$).

The pulse applied in the present study is similar to the one applied by Kemper et al. (Kemper et al., 2011) on 5 male volunteers close to 50th percentile submitted to sled test. Average corridor of mid-sternal compressions are compared to those obtained in this study (Kemper et al., 2011) on relaxed subjects ($n=5$) in Fig. 6. The average C_{max} are similar in both studies (7.9% Vs. 8%) even if midsternal compressions in the present study rose and dropped slower due to the difference in deceleration pulse characteristics, (4 g, 8 kph Vs. 5 g, 9.7 kph).

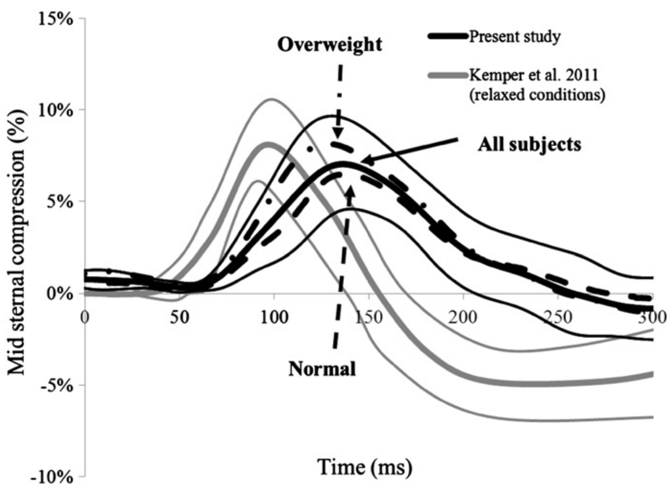

Fig. 6. Comparison of midsternal compression corridors from the present study and Kemper et al. (2011).

3.4. Force-deflection characteristics

Fig. 7 shows the relationship between the resultant shoulder belt load FRes and the mid-sternal deflection D for the normal group (a) and the overweightgroup (b). The overweight subject thorax was more compliant at the beginning of the loading and FRes increases with deflection continuing on up to D_{max}. There are no significant differences between overweight subjects (BMI ≥ 25 kg/m^2, $n=5$) and the subjects in the normal group (BMI <25 kg/m^2, $n=8$) in D_{max} (p=0.17) and as written before, $FRes_{max}$ was significantly higher for the overweight group than for normal group.

The average value of the dynamic stiffness (DS) was 54.8 ± 15.7 N/mm ($n=13$), greater than the average value of the effective stiffness (ES), 23.6 ± 10.1 N/mm ($n=13$). The different stiffness values are given in Table 2. Average ES was 30.9 ± 10.6 N/mm ($n=5$) for overweight subjects, significantly higher ($p<0.02$) than 19.6 ± 8.9 N/mm ($n=5$) for normal group, Fig. 8a. Inversely, average DS was 42.7 ± 8.71 N/mm ($n=5$) for the overweight group, significantly lower ($p<0.05$) than 61.7 ± 15.5 N/mm ($n=5$) for the normal group, Fig. 8b. As only one elderly subject is available in the present study, no assumption can be made about the influence of age on thoracic stiffness.

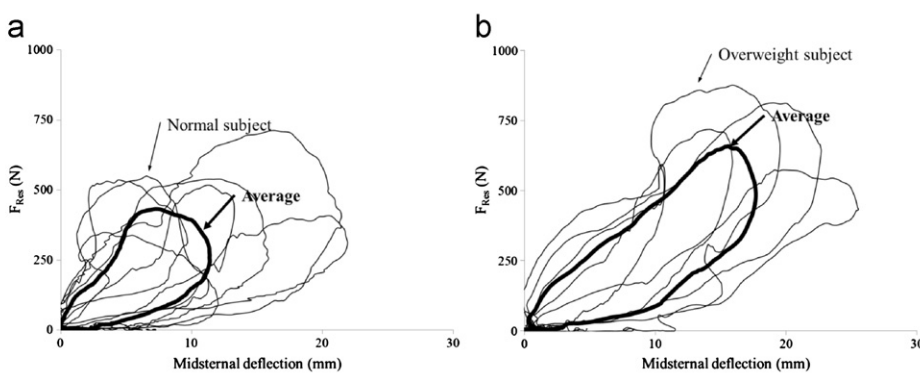

Fig. 7. Force-deflection of thorax volunteers for normal group (a) and overweight group (b).

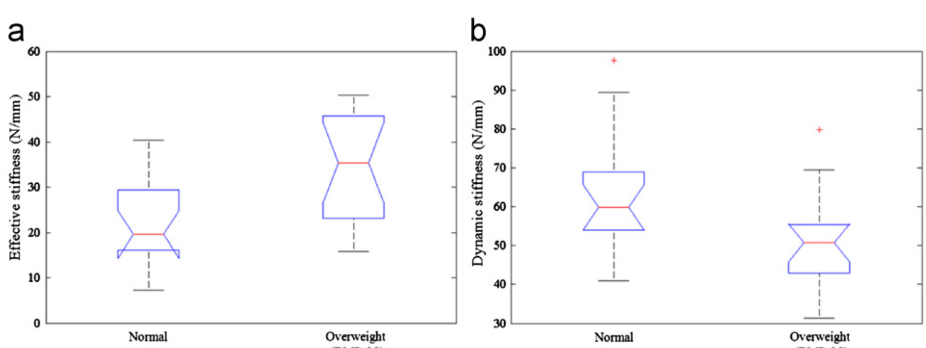

Fig. 8. Box plots of the effective stiffness (a) and dynamic stiffness (b) in N/mm for normal subjects and overweight subjects.

4. Discussion

13 volunteers of different anthropometries, genders and age were submitted to a low deceleration pulse. Thoracic mechanical response quantified using non-invasive techniques was linked to individual characteristics and compared to literature. In particular, the influence of overweight on thoracic mechanical response was evaluated.

Thoracic mechanical responses obtained in the present protocol are similar to those obtained in sled tests (Kemper et al., 2011). The variability of thoracic mechanical response is higher in the present study than in Kemper et al. (2011) where only 50th percentile males were considered.

The overweight volunteer group shows significantly higher ES and lower DS than the normal group. Increased compliance due to increased superficial tissue depth may cause the decrease in DS observed for overweight subjects at the beginning of the load. Since no significant difference was observed in D_{max} in both groups, the $FRes_{max}$ difference due to inertia, should explain consequently the higher ES observed for overweight subjects. Contrary to the trend highlighted by (Forman et al., 2009), no significant difference was found in C_{max} for overweight group or for the few obese subjects ($BMI \geq 30\ kg/m^2$, $n=1$) available in the present study.

As $FRes_{max}$ was also significantly linked to individual characteristics while $FB3_{max}$ not, the implementation of loading direction in the computation of belt force resultant is important. Although FRes is a 2D simplified computation, this approximation seems very adequate for the purposes of the current study and is easily reproducible for further studies.

Volunteer experiments overcome limitations related to selection criteria of human surrogates but have to face other challenges such as impact severity far below crash conditions and lower measurement accuracy due to the use of non-invasive techniques. Despite these limitations, this study could help in investigating the role of BMI in thoracic mechanical response variability and in assessing the thoracic mechanical response of a specific population, such the overweight and obese.

To the authors' knowledge, this is the first study to compare thoracic mechanical responses from overweight subjects to normal subjects using frontal sled tests with volunteers. The overweight subjects exhibited significantly greater resultant belt forces, higher effective stiffness and lower dynamic stiffness. Increased soft tissue depth and increased thorax may explain these observed differences.

The recruitment of volunteers has been continued including the elderly. It should allow isolating the influence of age on thoracic mechanical response independently from anthropometrical variations.

Conflict of interest statement

This study was partly funded by Toyota Motor Europe. S. Compigne is employee of Toyota Motor Europe.

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2012.12.021.

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