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

David Poulard [a,*], François Bermond [a], Sabine Compigne [b], Karine Bruyère [a]

[a] Université de Lyon, F-69622, Lyon; IFSTTAR, LBMC, UMR_T9406, F-69675, Bron, Université Lyon 1, Villeurbanne, France
[b] Toyota Motor Europe, Zaventem, Belgium

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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

* Correspondence to: LBMC—Site Ifsttar Lyon-Bron, 25 Avenue, Mitterrand, Case 24, 69675 Bron Cedex, France. Tel.: +33 478656873; fax: +33 472376837.
E-mail addresses: david.poulard@gmail.com, david.poulard@ifsttar.fr (D. Poulard).

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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*