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Biomechanical analysis of thorax-abdomen response of vehicle occupant under seat belt load considering different frontal crash pulses

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Purpose: The purpose of this work was to understand the biomechanical response and injury risk of thorax and abdomen of vehicle front seat occupants caused by seat belt load under different frontal crash pulses. *Methods*: A vehicle-seat-occupant subsystem finite element (FE) model was developed using the a assembly of vehicle front seat and seat belt together with the THUMS (Total Human body Model for Safety) AM50 (50^{th}% Adult Male) occupant model. Then the typical vehicle frontal crash pulses from different impact scenarios were applied to the vehicle-seat-occupant subsystem FE model, and the predictions from the occupant model were analyzed. *Results*: The modeling results indicate that the maximum sternal compression of the occupant caused by seat belt load is not sensitive to the peek of the crash pulse but sensitive to the energy contained by the crash pulse in the phrase before seat belt load reaching its limit. Injury risk analysis implies that seat belt load of the four crash scenarios considered in the current work could induce a high thorax AIS2+ injury risk (>80%) to the occupants older than 70 years, and a potential injury risk to the spleen. *Conclusions*: The findings suggest that control of the energy in the first 75 ms of the crash pulse is crucial for vehicle safety design, and thorax tolerance of the older population and spleen injury prevention are the key considerations in developing of seat belt system.

Key words: *thorax and abdomen, biomechanical response, injury risk, seat belt load, frontal crash*

1. Introduction

Accident analysis indicated that the excessive seat belt load in vehicle frontal crashes is a serious threat to the safety of occupants as which usually lead to thorax skeletal fractures and injuries to abdomen organs [5], [17], [18], though the improved design of load limiter in the seat belt system have largely reduced the risk of occupant injury from seat belt load in crashes, further developments of seat belt system is still in need.

In the past decades, many studies have focused on analysis of vehicle occupant injury risk from seat belt loading based on accident data analysis, numerical simulations and crash tests. For example, accident data analysis found that the major cause of chest injury to occupants in real world vehicle frontal impacts was the restraining loads from the seat belt [5]. Modeling work on the correlation among seat belt load and occupant chest response indicated that seat belt load limit, seat belt type and the the combination of the seat belt and airbag load could affect occupant thorax injury risk [9]. Some studies proposed adapting load limiter deployment and investigated how varying load limiter thresholds could affect occupant kinematics and injury outcome in frontal impacts based on simulations using multi-body models of Hybrid III dummies, and found that the optimal belt load limit varies from impact scenarios and occupant sizes [4], [6]. All the above studies indicated that seal belt load is a key factor affecting vehicle occupant safety in frontal crashes. How-

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ever, biomechanical analysis of occupant thorax and abdomen response due to seat belt load considering different crash scenarios is scarce, further improvement of seat belt system design still lacks fundamental reference as real world vehicle frontal crashes could occur in various severity. On the other hand, finite element (FE) human body models are widely used in vehicle safety [2], [7], [9], [13], [15], [26], which can predict more detailed biomechanical response than the multi-body human body models and FE dummy models. Biomechanical analysis of occupant response due to seat belt load using FE human body models would provide additional detailed perspective to the existing findings.

Therefore, the purpose of the current study was to analyze the biomechanical response and injury risk of thorax and abdomen of vehicle front seat occupant caused by seat belt load under different frontal crash scenarios via FE simulations using the THUMS (Total Human body Model for Safety) AM50 (50th% Adult Male) occupant model (version 4.0) [24]. To achieve this goal, typical vehicle frontal crash scenarios such as full-width rigid wall impact, offset deformable barrier impact, truck rear under-run impact, and front-to-side impact were considered, and the predicted biomechanical response of skeleton and internal organs and injury risk from the thorax-abdomen segment of the THUMS model were analyzed.

2. Materials and methods

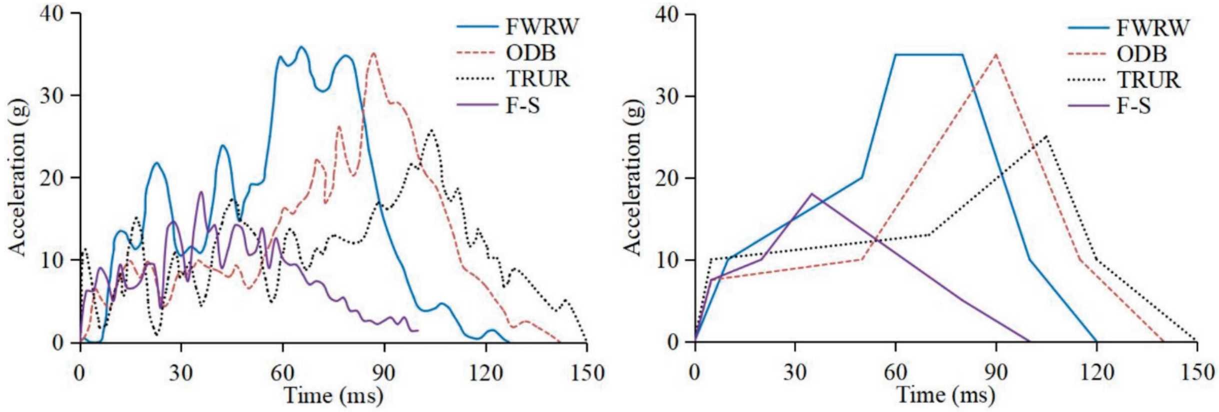
2.1. Vehicle–seat–occupant subsystem FE model

To simulate the interaction between the occupant and vehicle seat and seat belt system, a vehicle-seat-occupant subsystem FE model was developed using the assembly of a vehicle front seat, seat belt, floor and side frame together with the THUMS AM50 occupant model (Fig. 1). The use of this subsystem model can reduce computing time, similar approach was widely used in analysis of occupant safety regardless of vehicle body deformation [9], [15], [26]. The seat and seat belt system, including the floor and side wall structures, was extracted from the full size FE model of Honda Accord downloaded from the NHTSA (National Highway Traffic Safety Administration) website (https://www.nhtsa.gov/research-data/crash worthiness), which was validated against real car full frontal impact and oblique impact tests [23]. The seat belt model can simulate the characteristics of seat belt webbing elongation, retractor pay-out and load-limiter and retractor pretensioner [23], in the current study, the pretensioner load and load limit were set at 1.5 kN and 4.5 kN, respectively, and the restraint system was set to be activated at 0.014 seconds after the collision by sensors. The THUMS AM50 model has been validated against cadaver tests under different impact loading and applied in predicting real world injuries, where the biofidelity of this human body model has been widely recognized [7], [9], [13]. The friction coefficients of 0.2 and 0.35 were defined for the seat belt-to-occupant contact and occupant-to-seat contact, respectively.

Fig. 1. Vehicle–seat–occupant subsystem FE model

2.2. Vehicle frontal crash pulses

To assess the common seat belt loading to the vehicle occupants, the typical vehicle frontal crash pulses in the longitudinal direction (acceleration-time histories, Fig. 2a) from tests of full-width rigid wall (FWRW) impact (56 km/h), offset deformable (ODB) impact (64 km/h) and truck rear under-run (TRUR) impact (48 km/h), and from a real world front-to-side (F-S) impact accident case were considered. The crash pulses of FWRW and ODB impacts were sourced from the test data shared by the NHTSA [18], [23], and the curve for TRUR impact was from the study by Rechnitzer et al. [20]. The real world accident case is from the database of the Institute for Traffic Medicine, Army Military University, China, where the target vehicle



(a) (b)

Fig. 2. Crash pulses for different impact scenarios: (a) the original pulses and (b) simplified pulses

Fig. 3. The crash scenario of the real world front-to-side impact case

(white) front impacted with the side of another car (black) at the speed of 28 km/h (Fig. 3) and the crash pulse of the target vehicle was recorded by the Crash Data Retrieval (CDR) system.

In order to the compare the characteristics of different crash pulses clearly, the original curves were adapted into simplified pulses which can basically capture the trend and magnitude of the original crash pulses, where the peak value and knee points were used to build the simplified curve (Fig. 2b). It can be observed that the crash pulse for FWRW impact appears a "double trapezoid" characteristic, while the crash pulses for ODB, TRUR and F-S impacts show the shape of "trapezoid + triangle". For the selected four crash pulses in the current study, the impact energy induced by the FWRW impact is highest, followed by the ODB and TRUR impact, and the F-S impact is most moderate, which can generally cover the range of impact severity observed in real world accidents [4]. These simplified frontal crash pulses were then applied to the vehicle body parts of the seat–occupant subsystem for simulations conducted by LS-DYNA software [14], which can induce accelerations to the seat and seat belt assembly and hence a pull loading to the occupant model.

2.3. Injury assessment metrics

To assess the potential injury risk of vehicle occupant thorax and abdomen under seat belt load, the predicted biomechanical response, including sternal and abdominal compression and stress of the solid organs, from the THUMS model in the simulations were analyzed and compared between different crash scenarios. These selected biomechanical metrics are regarded as the key indexes for thoracic and abdominal injuries [1], [9]–[12], [22]. In Figure 4, the particular measuring approach (red arrows) for sternal and abdominal compression is shown. The sternal compression was defined as the change of relative distance between the outer surface of the sternum and the spine surface, and two measuring points (upper and middle-upper sternum) were set with the maximum data being used for analysis. The abdominal compression was measured from the change of relative distance between the intraperitoneal wall (at the location of seat belt contact) and the spine surface. Both sternal and abdominal compression were measured at the sagittal plane.

For quantity estimation of thorax injury risk, the AIS2+ thorax injury risk curve developed by Ekambaram et al. [4] according to the cadaver test data summarized in the study of Laituri et al. [12] was employed, where the probability of age-dependent AIS2+ thorax injury was defined as a function of thorax deformation for crash test dummies. Considering the difference between dummies and human body [12], in the current study this injury risk curve was adapted as a function of normalized thorax compression (NTC: compression divided by its initial depth) based on the relationship between thorax deflection of dummies and normalized thorax compression (NTC) of cadavers validated by Laituri et al. [12]. The AIS2+ thorax injury risk as a function of NTC is given as below:

$$p(\text{AIS2+}) = \frac{1}{1+\text{EXP}(12.432-0.0562*\text{Age}-30.866*\text{NTC})}. \quad (1)$$

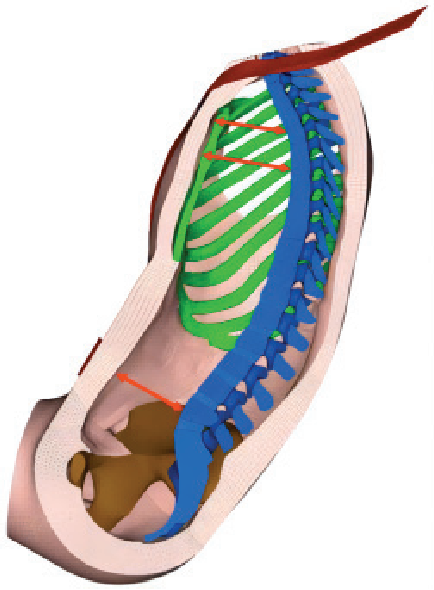


Fig. 4. Measuring approach of sternal and abdominal compression

3. Results

3.1. Thorax and abdomen compression

The predicted overall compression process of the thorax and abdomen for the occupant model under the

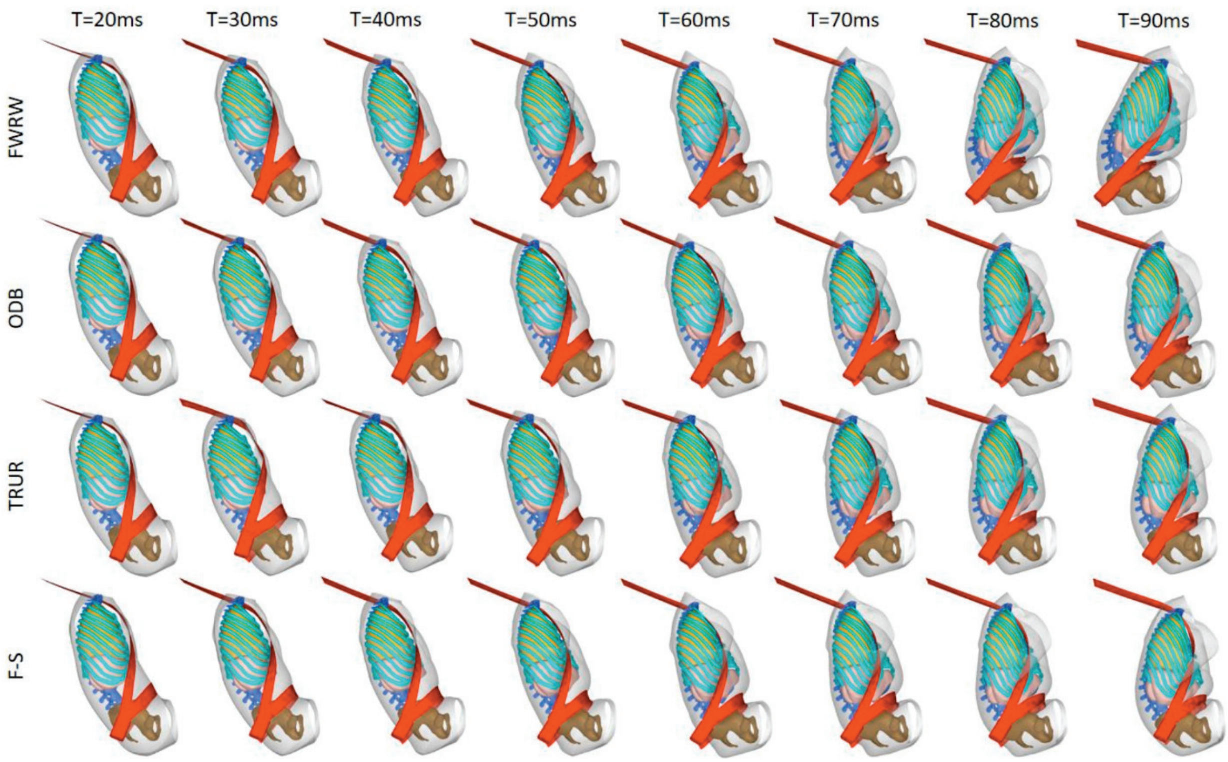


Fig. 5. Overall compression process of the thorax and abdomen for different crash scenarios

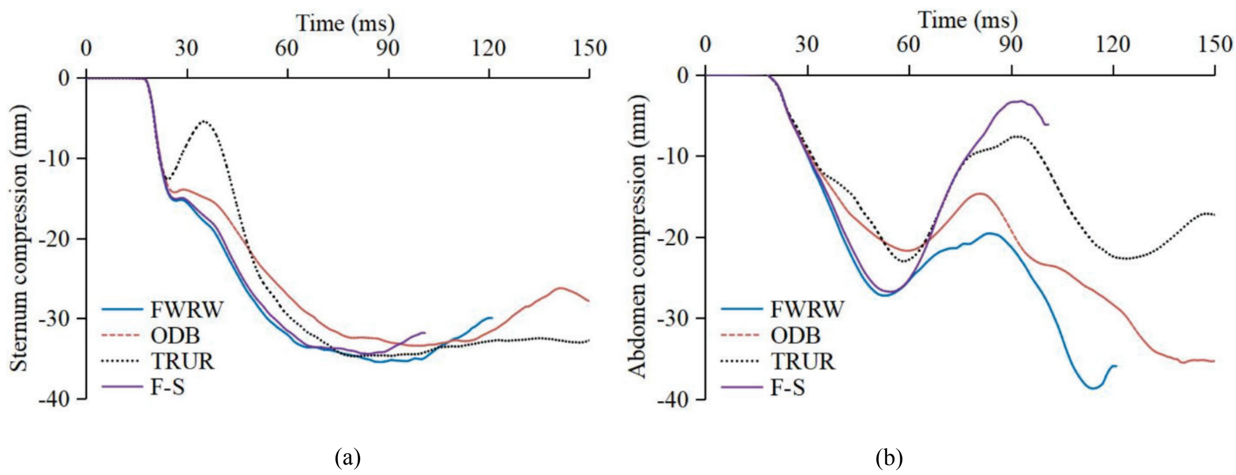


Fig. 6. Time history curves for compression of the sternum (a) and abdomen (b)

seat belt load during the simulations for different vehicle frontal crash pulses is shown in Fig. 5, and the time history curves for compression of the sternum (upper measuring point) and abdomen are plotted in Fig. 6. It is observed that the thorax compression starts at about 20 ms after loading, and reaches the maximum value of around 35 mm at the period of 60–90 ms after loading. Generally, there is no significant difference in maximum sternal compression between different impact scenarios, but the FWRW and ODB crashes have a obviously higher abdominal compression at the second peak (nearly 40 cm).

3.2. Injury risk estimation

The estimated AIS2+ thorax injury risks for different impact scenarios from Eq. (1) are shown in Table 1. Since the thorax injury risk is age-dependent, three age groups were assumed in the estimation of AIS 2+ thorax injury risk, which was set as 30, 50, and 70 years for young, middle-aged and older occupants, respectively, according to the statistic data from a previous study [4]. The FWRW impact has the highest AIS2+ thorax injury risk to the occupants, which reaches 64.8% and 85% for the middle-aged and older occupants. The ODB impact has the lowest AIS2+ thorax injury risk, where only the older occupants have a risk over 50%.

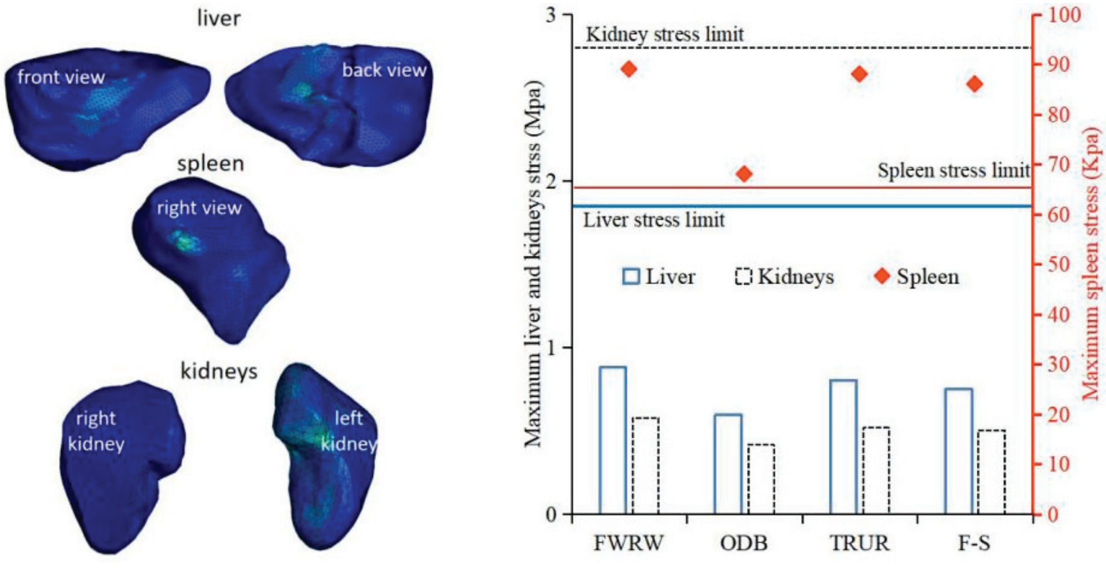
Table 1. Estimated thorax injury risk for different impact scenarios

Scenario		FWRW	ODB	TRUR	F-S
NTC		0.332	0.312	0.325	0.322
AIS2+ thorax injury risk	Young	37.4%	24.7%	32.9%	31.5%
	Middle-aged	64.8%	50.2%	60.1%	58.6%
	Older	85.0%	75.6%	82.3%	81.3%

For the abdomen injury estimation, no injury risk curves could be refereed from the literature, the predicted stress of the internal organs and the average ultimate stress (referring as the thresholds) from experimental data [3], [8], [27] were employed for injury risk estimation. The predicted typical stress distribution of the parenchyma and capsule tissue of the liver, spleen and kidneys, and the maximum stress of these organs are shown in Fig. 7 together with the thresholds. High stress was observed in the front and back surface in the middle area of the liver, right side wall of the spleen and left kidney (Fig. 7a). Compared to the ultimate stress references, the predicted maximum stress for the liver (<1 MPa) and kidneys (<0.6 MPa) are far lower than the thresholds (blue = 1.85 MPa and black = 2.8 MPa lines), while the maximum of the spleen is higher than the threshold (red line = 65 kPa) (Fig. 7b). On the other hand, the FWRW impact has a higher maximum stress to the liver, spleen and kidneys than other crashes, while the ODB crash has a lower maximum stress for all the above internal organs than other scenarios.

4. Discussion

In vehicle frontal crashes, seat belt load is an important source for occupant thorax and abdomen injuries. The current work analyzed vehicle occupant thorax-abdomen biomechanical response and injury risk under seat belt load in different frontal crash pulses using a human body FE model with high biofidelity. The simulation results show significant compression to occupant thorax-abdomen segment (Figs. 5, 6), the maximum sternal compression is about 35 mm which is similar to that reported in the literature [9]. Com-



(a) (b)

Fig. 7. Typical distribution (a) and the maximum stress (b) of the liver, kidneys and spleen

bining the simulation results with the input data in Fig. 2, it is found that the maximum sternal compression seems to be not very sensitive to the peek of the crash pulse but sensitive to the energy contained by the first 70 ms of the crash pulse, e.g., the maximum sternal compression of the ODB case (peak pulse = 35 g) is lower than the F-S impact (peak pulse = 18 g). This is mainly because the simulated characteristics of retractor pay-out and load-limiter (4.5 kN) avoid overload to the occupant thorax, which allows only the load at first stage (before 75 ms) to be transmitted to the occupant thorax, but the abdominal strap still restricts (Fig. 8). Clearly, the shoulder belt (section A) load reaches its limit at 75 ms and keeps this level until the unloading phase. The load limiting effect of the shoulder belt can also be observed from the dynamic response of the occupant in Fig. 5 where the torso of the occupant tends to rotate forward after 70 ms of the simulations. Since the impact energy contained in the first 75 ms of the ODB crash pulse is relatively lower than other scenarios (Fig. 3), the predicted maximum sternal compression (Fig. 6a) and maximum stress of the upper abdomen organs (Fig. 7b) from the ODB crash are lower than other impacts. This observed trend of the crash pulse affecting thorax compression is similar to that reported in the literature [6], where the peak of crash pulse is also not strictly associated with the thorax compression of dummy models. However, the abdomen belt continues to be loaded as the torso leans forward, which leads to a second peak to the seat belt force (Fig. 8) and also the abdomen compression (Fig. 6b). Considering the abdomen compression was measured from the lower abdomen, which mainly compresses the cavitary organs (large and small intestine) but induces no direct compression to the upper abdominal organs, the maximum stress of the liver, kidneys and spleen (Fig. 7b) are hence not associated with the maximum abdomen compression (Fig 6b). The above findings work may suggest that control of the energy in the first 75 ms of the crash pulse and seat belt load limit is crucial for vehicle structure and seat belt design. However, it is still a challenge to design an optimal seat belt load limiter for all impact scenarios and occupants [4], [6], [9] and there might be adverse effects from belt load limiters considering the increases in excursion and head contact risk [2].

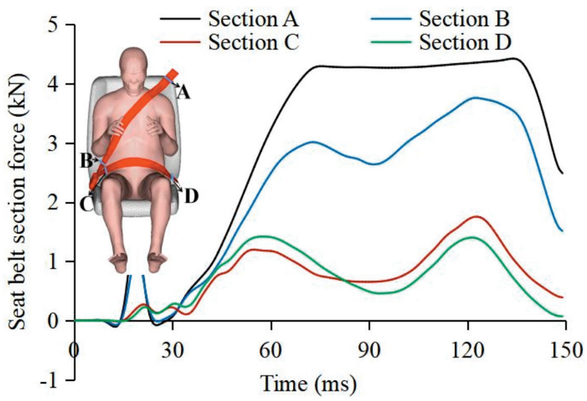


Fig. 8. Seat belt section force time history for the TRUR impact

The injury risk estimation results indicate that older occupants may have a high AIS2+ thorax injury risk (>75%) under seat belt load for all crash pulses. A previous study based on simulations using a dummy model and the seat belt system of 4 kN load limit also found a high thorax AIS2+ injury risk to older occupants [4]. This finding implies that the thorax tolerance of the older population is the key consideration in developing seat belt system, which was also emphasized in previous studies [4]–[6], [10]–[12]. On the other hand, since the thorax injury risk is highly sensitive to the normalized thorax compression (Eq. (1)), e.g., an obvious difference (>10%) in thorax AIS2+ injury risk was observed between the ODB crash and FWRW impact for a small gap (0.02) in normalized thorax compression, a small improvement in seat belt design or vehicle crash pulse for reducing thorax compression could play an important role in protecting thorax injury. For the abdomen, quantitative injury risk is not able to be analyzed as no risk curves could be referred given the complex nature of the abdomen and the difficulties associated with the performance of adequate experiments. But the simulation results indicate that the predicted maximum stress upper abdomen organs is associated with the maximum thorax compression, e.g., the FWRW impact is the highest and the ODB crash is the lowest. The simulation results also show that the peak stress of the internal organs could be induced by the directly compression of the thorax (front surface of the liver and right side wall of the spleen) and also the mutual squeezing between the internal organs followed by thorax compression (back surface the liver and left kidney). Thus, thorax compression could be regarded as a index for assessing injury risk of upper abdomen organs, but further analysis is needed to develop a quantitative relationship. The predicted biomechanical response of the upper abdomen solid organs shows that the maximum stress of spleen is higher than the average ultimate stress observed from experimental data [8], which indicates a potential injury risk to vehicle occupant spleen under seat belt load. Many studies have also reported that the spleen is the most frequently injured solid organ in blunt abdominal trauma, where vehicle crashes are the most frequent cause of spleen injury [1], [19], [21]. Previous simulation work under seat belt load using the THUMS model also found that the spleen strain is over 0.8 and is higher than other organs [9]. This finding suggests that it is necessary to evaluate spleen injury risk in vehicle safety assessment, where further improvements to the current test dummies and relative injury criterion are in need.

There are several limitations to the current study. Firstly, only 4 crash pulses were employed, more scenarios could be considered, though the crash pulses can generally represent the population of impact loads in different severity reported in the literature [4]. Secondly, the contact of the occupant to the front airbag and other interior parts were not considered. However, this would not affect the finding of the current significantly, given the fact that the thorax compression reaches its maximum value at the time near the that of the airbag fully deployed [23]. Finally, the response of rear seat occupants was not studied in the current work, which is worth to analyze as seat belt has a significant influence on the safety of rear seat occupants [16]. Nevertheless, the current work provides additional detailed perspective to the existing understanding on vehicle occupant thorax and abdomen response under seat belt load.

5. Conclusions

The current study analyzed the biomechanical response and injury risk of thorax and abdomen of vehicle occupants under seat belt load from different frontal crash pulses via FE simulations using the THUMS model. The modeling results indicate that the maximum sternal compression of the occupant caused by seat belt load is not sensitive to the peek of the crash pulse but sensitive to the energy contained by the crash pulse in the phrase before seat belt load reaching its limit. Injury risk analysis implies that seat belt load of the four crash scenarios considered in the current work could induce a high thorax AIS2+ injury risk (>80%) to the occupants older than 70 years, and a potential injury risk to the spleen. The findings of the current work indicate that control of the energy in the first 75 ms of the crash pulse is crucial for vehicle safety design, and thorax tolerance of the older population and spleen injury prevention are the key considerations in developing of seat belt system, which may suggest that future seat belt system design should offer more focuses on protection of seniors and reducing thorax-abdomen compression with the help of adapting load limiter and airbelt as recommended by previous studies [4], [6], [9].

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