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Pelvis and thoracolumbar spine response in simulated under-body blast impacts and protective seat cushion design

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Purpose: The aim of this study is to investigate the dynamic and biomechanical response of the pelvis and thoracolumbar spine in simulated under-body blast (UBB) impacts and design of protective seat cushion for thoracolumbar spine injuries. *Methods*: A whole-body FE (finite element) human body model in the anthropometry of Chinese 50th% adult male (named as C-HBM) was validated against existing PHMS (Postmortem Human Subjects) test data and employed to understand the dynamic and biomechanical response of the pelvis and thoracolumbar spine from FE simulations of UBB impacts. Then, the protective capability of different seat cushion designs for UBB pelvis and thoracolumbar injury risk was compared based on the predictions of the C-HBM. *Results*: The predicted spinal accelerations from the C-HUM are almost within the PHMS corridors. UBB impact combined with the effects from physiological curve of the human thoracolumbar spine and torso inertia leads to thoracolumbar spine anterior bending and axial compression, which results in stress concentration in the segments of T4–T8, T12–L1 and L4–L5. Foam seat cushion can effectively reduce the risk of thoracolumbar spine injury of armored vehicle occupants in UBB impacts, and the DO3 foam has better protective performance than ordinary foam, the 60 mm thick DO3 foam could reduce pelvic acceleration peak and DRIz value by 52.8% and 17.2%, respectively. *Conclusions*: UBB spinal injury risk is sensitive to the input load level, but reducing the pelvic acceleration peak only is not enough for protection of spinal UBB injury risk, control of torso inertia effect would be much helpful.

Key words: *under-body blast, thoracolumbar spine, biomechanical modelling, protective cushion*

1. Introduction

Landmines and improvised explosive devices (IEDs) are the main anti-tank weapons in asymmetric warfare, which can release huge energy in explosion at the bottom of an armored vehicle, producing under body blast (UBB) loads [2]. The UBB load, large magnitude and short duration acceleration, poses a serious threat to the lower limbs, pelvis and thoracolumbar spine of armored vehicle occupants, where the thoracolumbar spinal fractures may cause spinal nerves and the spinal cord injuries, which could produce long-term or even permanent disability [3], [15], [17]. Therefore, it is of great significance to carry out research on understanding the mechanisms and protection approaches of armored vehicle occupant thoracolumbar spine injuries in UBB impacts.

Currently, PMHS (Postmortem Human Subjects), ATDs (Anthropomorphic Test Devices) and ATD numerical models are the main avenues in analysis of armored vehicle occupant UBB spinal injury and protection. For example, Yoganandan et al. conducted repetitive PMHS tests to study the mechanism of load transmission and potential variables affecting the injury risk of pelvis and spinal structures [21], and develop temporal corridors of loads at the pelvis and spine and assess clinical fracture patterns [22]. Bailey et al. [1] tested seven PMHS with a horizontal skid steer and analyzed the response and injury of the pel-

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vis in different acceleration ranges. Dooley et al. [4] loaded 23 PMHS thoracolumbar segments (T7–L5) at different rates and investigated the effect of loading rate on the degree of vertebral injury. Pandelani et al. [14] investigated the mechanism of pelvic fractures via axial impact tests using three fresh-frozen male pelvic specimens. Apart from these segment studies, whole-body PMHS UBB tests were also employed to assess the predictive capability of the Hybrid III dummy in representing the PMHS response [13], investigate more injurious in whole-body conditions [16], or understand the mechanisms and timing of spine injuries [18]. Recently, with the development of numerical modelling technology, finite element (FE) human body models developed based on human anatomical structure and biomaterials have been widely applied in the studies of impact injuries. Zhang et al. [25] simulated the response of lumbar spine under high-speed vertical load using a FE model of human lumbar spine-pelvis-femur segment. Weaver et al. [23] used the GHBMC (Global Human Body Models Consortium) 50th percentile male human body FE model to evaluate the pelvic response under body blast. Somasundaram et al. [19] validated the biofidelity of the GHBMC model in UBB impacts against PMHS test data. FE human body models provides a possibility to explore the biomechanical response of the spine in UBB impacts. Axial compression applied through the pelvis together with flexion moment of the torso are regarded as the main mechanisms of thoracolumbar spine UBB injuries, and the characteristics of the input UBB load have a significant influence on the location and time of the spine [19]. Though much important information is known from previous studies, there is still a lack of biomechanical understanding on thoracolumbar spine response and injury mechanisms of occupants in UBB impacts. On the other hand, blast-resistant vehicle body and seat are the current focuses of attention for UBB injury protection research [8], relatively few studies have been conducted on the blast protection properties of seat cushions.

Therefore, the purpose of this study is to investigate the dynamic and biomechanical response of the pelvis-spine in simulated under-body blast (UBB) impacts and design of protective seat cushion for thoracolumbar spine injuries. Firstly, a whole-body FE human body model in the anthropometry of Chinese 50th% adult male was validated against PHMS UBB test data. Then simulations in different UBB load levels were carried out using the validated human body model to understand the kinematics and dynamic and biomechanical response of occupant pelvis and thoracolumbar spine in UBB impacts. Finally, the protective capability of different seat cushion designs for UBB thoracolumbar injury risk was compared based on dynamic and biomechanical predictions of the human body model.

2. Materials and methods

2.1. Human body validation

A seated human body finite element model of Chinese 50th% adult male (Fig. 1a), named as C-HBM (Chines Human Body Model) was used in this study to simulate armored vehicle occupants. The C-HBM occupant model, containing 1403260 elements and 332892 nodes, was developed using LS-DYNA codes [9] based on the human body geometry extracted from CT and MRI data of a volunteer in the anthropometry of Chinese 50th% adult male, which includes detailed skeleton and softer tissues (brain, thoracic and abdominal

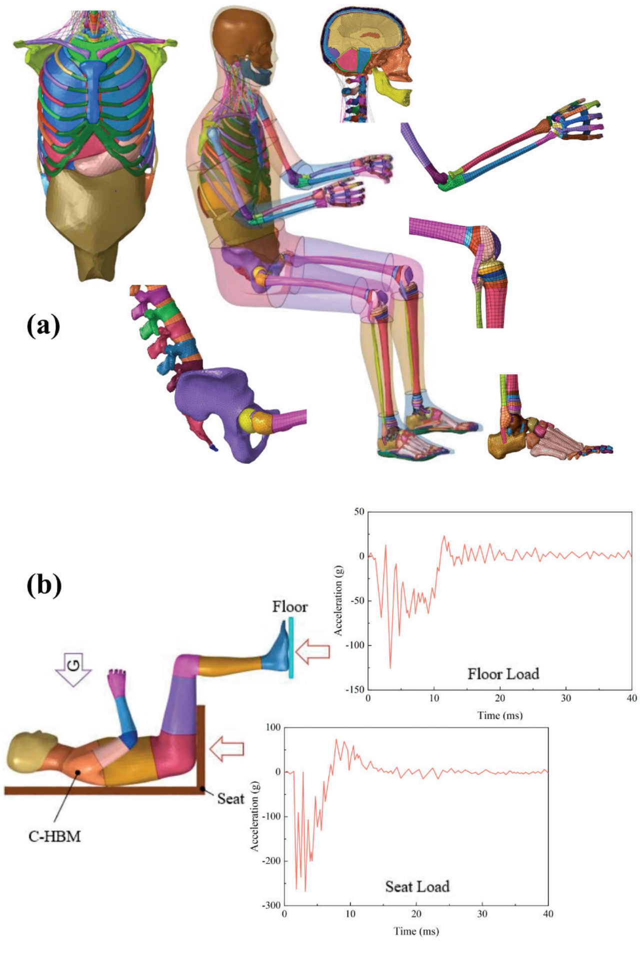

Fig. 1. C-HBM occupant model (a) and simulation model for the PMHS experiment (b)

organs, ligaments, skin, fats, muscles, etc.). The geometry of the Chinese 50th% adult male was initially constructed using Mimics software, which was then smoothed, amended and generated Nurbs surfaces with Geomagic studio software to create NURBS surfaces. In the FE human body modeling process, the Hypermesh software was employed to develop elements based on the geometry and define materials and properties. Hexahedral and tetrahedral elements were used for modelling the solid tissues (such as bones, organs, fats, and muscles), shell elements were employed for simulating ligaments and skin, 1-D elements were defined for setting muscle force. Material models such as elastic, elastic-plastic, viscoelastic, and Ogden were defined for different body parts in the C-HBM [11], [12]. Particularly, bone structures were modeled with separated regions concerning different properties and cortical bone thicknesses; organs were modeled using tetrahedral elements to fill the volume together with 1.0 mm thick shell elements to envelope the external surface of the viscera; skeletal muscles were built by hexagonal elements with detailed geometry and combined with 1D Hill-type beam elements of defined properties.

Since the current study focuses on occupant thoracolumbar spine response and injury protection, the biofidelity of the foot-leg-pelvis-spine segment of the C-HBM should be validated. The lower limb model of the C-HBM has been validated against PMHS UBB test data [5], and have been applied for biomechanical analysis in various impact loads [6], [10]. Therefore, the current work firstly validated the biofidelity of the C-HBM occupant model focusing on pelvis and spine response in UBB impacts to ensure the effectiveness of the findings. Particularly, the PMHS test conducted in the literature [18], [19] was simulated using the C-HBM, and the spinal dynamic response of the C-HMB was then compared with the PMHS test data for model validation. In PMHS tests, cadavers of adult male in the height of 177–184 cm were employed, where the angle of hip, knee and foot joint was set as 90° [18], [19]. In Fig. 1b, the simulation model for the PMHS test is shown, where the same posture of the cadaver and same acceleration pulses to the floor and seat were defined as the PMHS test.

2.2. Simulation matrix definition

Two simulation matrices were defined in the current work, the first one was set to understand the dynamic and biomechanical response of pelvis and thora-

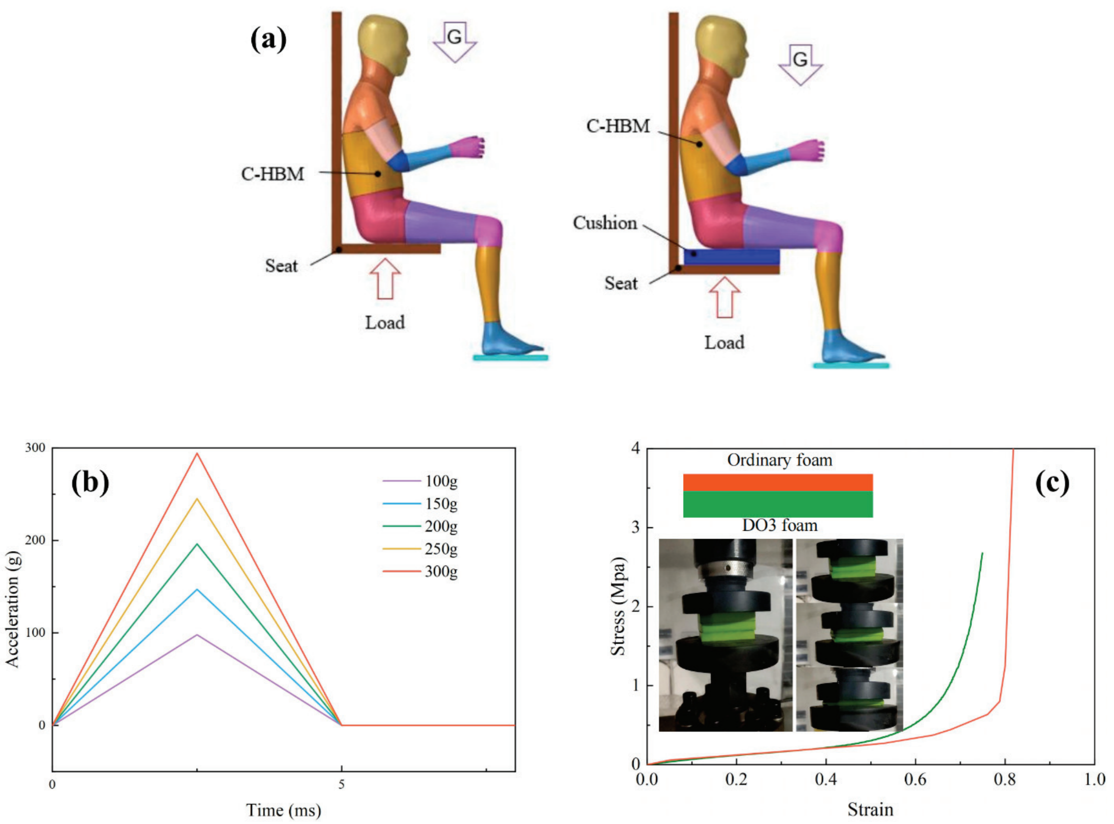

Fig. 2. Simulation models of UBB impacts without/with cushion protection using the C-HBM (a), UBB pulses with different energy levels (b), and stress–strain curves of ordinary and DO3 foam (c)

columbar spine in UBB impacts with different energy levels, another one was for comparing the protective capability of different seat cushion designs for thoracolumbar spine UBB injuries. In Figure 2a, the simulation models of the C-HBM occupant model in UBB impacts without/with cushion protection are shown, where a foam cushion in thickness of 60 mm was placed on the surface of the sea for the cases with protection. The thickness of the seat cushion was defined according to the dimension of a real product in developing. In Figure 2b, the input pulses of different energy levels are shown, where simplified 5 ms width triangle pulses with a peak of 100–300 g at 2.5 ms were employed according to previous analysis on typical UBB with different TNT equivalents [4]. The the material properties of the foam (obtained by quasi-static compression tests) for cushion design were shown in Fig. 2c, where the DO3 foam and ordinary polyethylene foam were used to design single material (DO3 or ordinary foam, 60 mm in thickness) cushions and combined material (DO3 + ordinary foam) cushions with different thickness combinations of DO3 and ordinary foam (DO3 + ordinary = 10 mm + 50 mm, 20 mm + 40 mm, 30 mm + 30 mm, 40 mm + 20 mm or 50 mm + 10 mm). It should be noted that only the moderate impact load of 150 g was used in the simulations with cushion protection, considering the attenuation of the UBB shock wave after the blast-resistant seat, and the purpose here is only to compare the protective capability of different cushion designs. In total ten simulations (five for dynamic response analysis in different UBB loads and five for protective capability comparison between different cushion designs) were conducted in the LS-DYNA software environment.

2.3. Data analysis

The overall kinematics of the occupant, pelvis acceleration and DRIz (Dynamic Response Index in the vertical direction) and von Mises stress in the cortical bone of the thoracolumbar spine were used to analyze armored vehicle occupant thoracolumbar spine response in UBB, while in the study of protective capability of cushion design, the normalized values (referring to the unprotected case) of these parameters were applied for a relative comparison.

The DRI was proposed to evaluate the injury possibility of human thoracolumbar vertebrae under the action of axial impact force [20]. The DRI theoretical model simplifies the human thoracolumbar spine as a single-mass spring damper system, with the pelvic axial acceleration as the input to this system, and the value of the DRI is derived from the maximum relative displacement calculated from this system. The equation of the DRI model is given as:

$$\frac{d^2z}{dt^2} = \frac{d^2\delta}{dt^2} + 2\zeta\omega_n\frac{d\delta}{dt} + \omega_n^2\delta, \tag{1}$$

where: $\frac{d^2z}{dt^2}$ is the acceleration in the vertical direction of the pelvis; δ is the relative displacement of the system; ζ is the damping coefficient with a value of 0.224; ω_n is the intrinsic frequency with a value of 52.9 rad/s. DRIz indicates the DRI value in the vertical direction, Calculated from maximum relative displacement $\delta_{\max}$, ω_n and gravity acceleration:

$$\text{DRI}z = \frac{\omega_n\delta_{\max}}{g}. \tag{2}$$

3. Results

3.1. Validation of the C-HBM

In Figure 3, the acceleration curves on the thoracic spine (T5, T8, T12) and sacrum (S1) output from the C-HBM occupant model compared with the PHMS test corridors are shown. The predicted curves are generally in the similar trend as the test data and the simulation results almost lie within the range of the test corridors, which implies that the C-HBM occupant model can basically predict the response of human spine and sacrum in the UBB condition.

3.2. Dynamic and biomechanical response

In Figure 4, the typical kinematics of the occupant in UBB impacts are shown, which reflects the loading process where the energy of the seat is instantaneously transferred to the hip, then to the sacrum through the articular cartilage, to the spine (lumbar, thoracic, and cervical vertebrae), and finally to the head. The upper body stars with thoracolumbar spine compression and forward bending, followed by backward bending of the cervical spine, and finally the spine rebounds. The maximum compression of the spine occurs during the period about 20–30 ms after the impact, which also induces compression to the ribcage and abdomen.

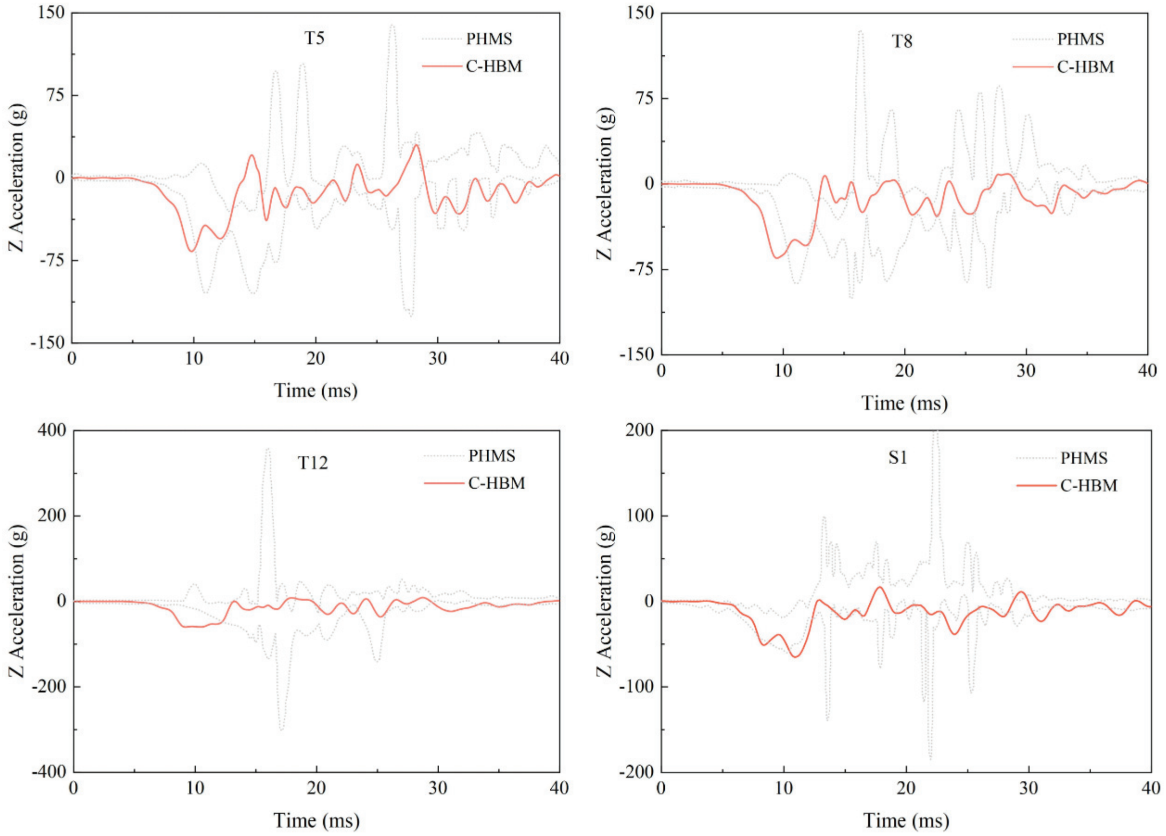

Fig. 3. Comparison of predicted spinal and lumbosacral acceleration and the PHMS test corridors

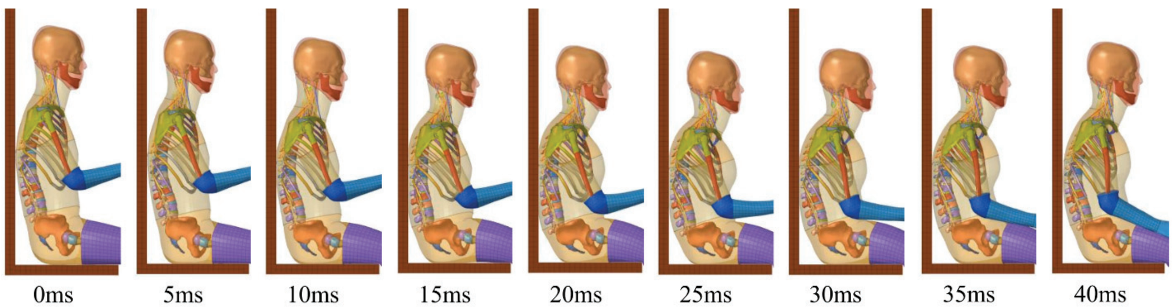

Fig. 4. Typical kinematics of the occupant model in simulated UBB impacts

The acceleration and DRIz time history curves of the pelvis in different UBB loads are shown in Fig. 5. The peak pelvic acceleration increases as the impact energy becomes larger, and the pelvic acceleration reaches the peak between 5–10 ms after the impact, then gradually decreases and a second peak appears at around 20 ms after the impact, and slowly returns to the baseline. The maximum pelvis acceleration floats from 50 to 150 g when the UBB loads changes from 100 to 300 g. The DRIz values corresponding to the 100–300 g UBB loads from are 8.8, 15.1, 20.1, 26.2, and 31.5, respectively. It is obvious that the DRIz value increases with the increase of the input UBB load, and the DRIz values at the UBB loads of 200–300 g exceeded the threshold value of 17.7, which indicates a risk of thoracolumbar spine injury according to the threshold [20].

In Figure 6a, the peak values of pelvis and thoracolumbar von Mises stress in different UBB loads are compared, where the pelvis peak von Mises stress generally increases with increasing UBB load and the peak thoracolumbar von Mises stress is 115 MPa in 100 g UUB load and floats from 210 to 225 MPa in the cases of 150–300 g. The distribution of the pelvis and thoracolumbar spine von Mises stress for the cases of 150 g and 250 g UBB loads (taking as examples), respectively, are shown in Fig. 6b. During the UBB impact, the stress concentration of the pelvis lies in the

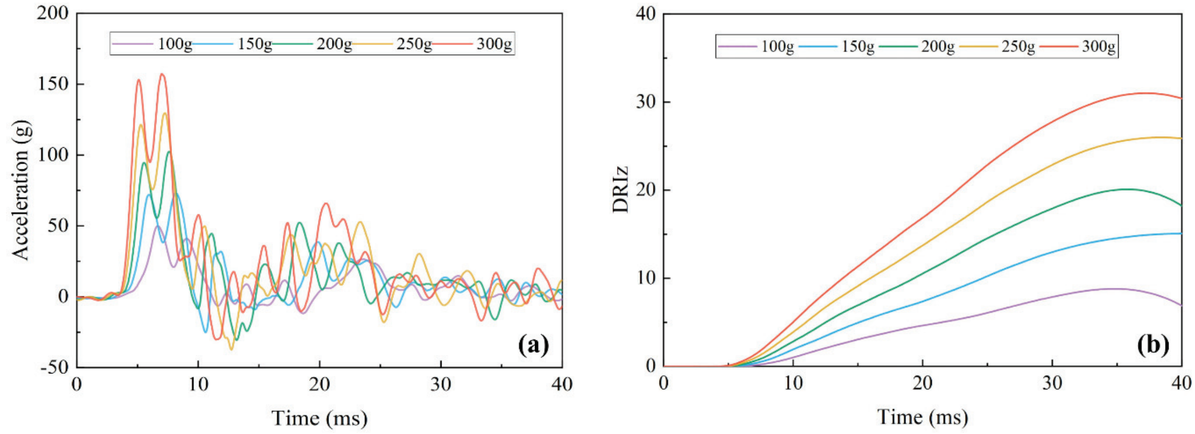

Fig. 5. Pelvic acceleration (a) and DRIz (b) response under different UBB impacts

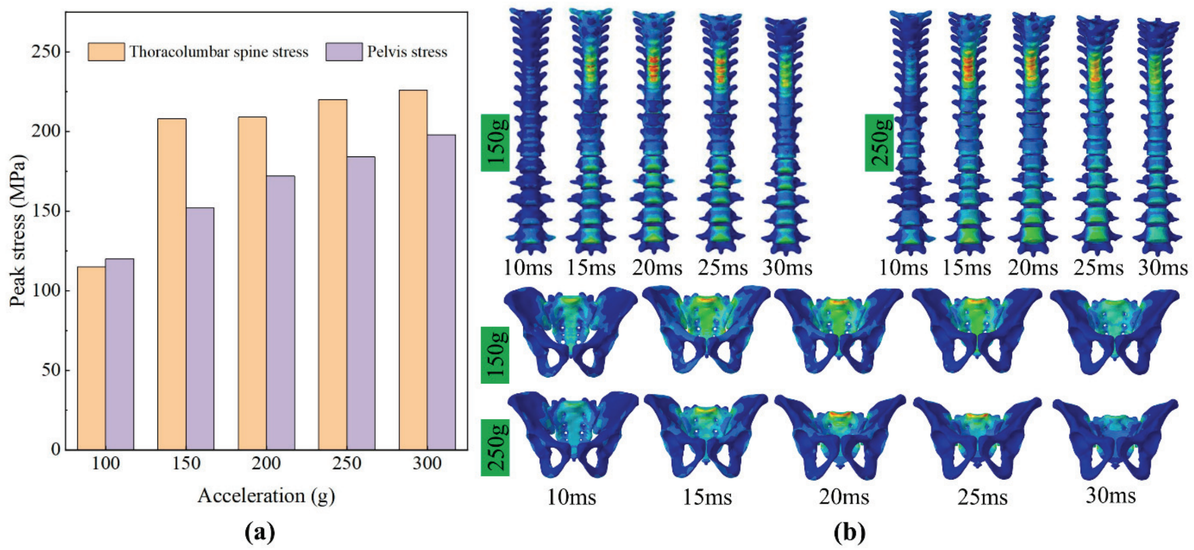

Fig. 6. Pelvic and thoracolumbar von Mises stress peak (a) and distribution (b) in different UBB impacts

iliac-sacral joint and the anterior part of the sacral-lumbar joint, while stress in the thoracolumbar spine is mainly concentrated in the segments of T4–T8, T12–L1, and L4–L5. The stress peak time and duration in the thoracolumbar spine are sensitive to the UBB load level, where the stress in the 150 g UBB load peaks from 20 to 25 ms and this is in the time range from 15 to 25 ms for the 250 g UBB load case.

3.3. Protective capability

In Figure 7a, the maximum thoracolumbar spine von Mises stress, pelvic acceleration and DRIz value predicted from the C-HBM occupant model in the cases without cushion protection and with the protection of different cushions are compared. The cushion has a significant effect on the reduction of the peak pelvic acceleration, but a moderate and weak influence on DRIz value and thoracolumbar spine stress peak, respectively. The peak pelvic acceleration is minimized with the protection of a cushion of 50 mm + 10 mm for the combination thickness of ordinary foam + DO3 foam, which is 54.4% lower than that without protection. The minimum DRIz value occurs in the case protected by a cushion of pure DO3 foam, where the peak pelvic acceleration, thoracolumbar spine von Mises stress and DRIz reduced by 52.8, 4.7 and 17.2% compared to the case without cushion protection, respectively. In addition, the peak pelvic acceleration decreases with the increase of the thickness of the DO3 foam layer in the combined cushion, but the DRIz value and maximum thoracolumbar von Mises stress have no gradually decreasing trend with increasing the thickness of the DO3 foam layer. However, the area and duration of high stress concentration in the thorax spine decrease

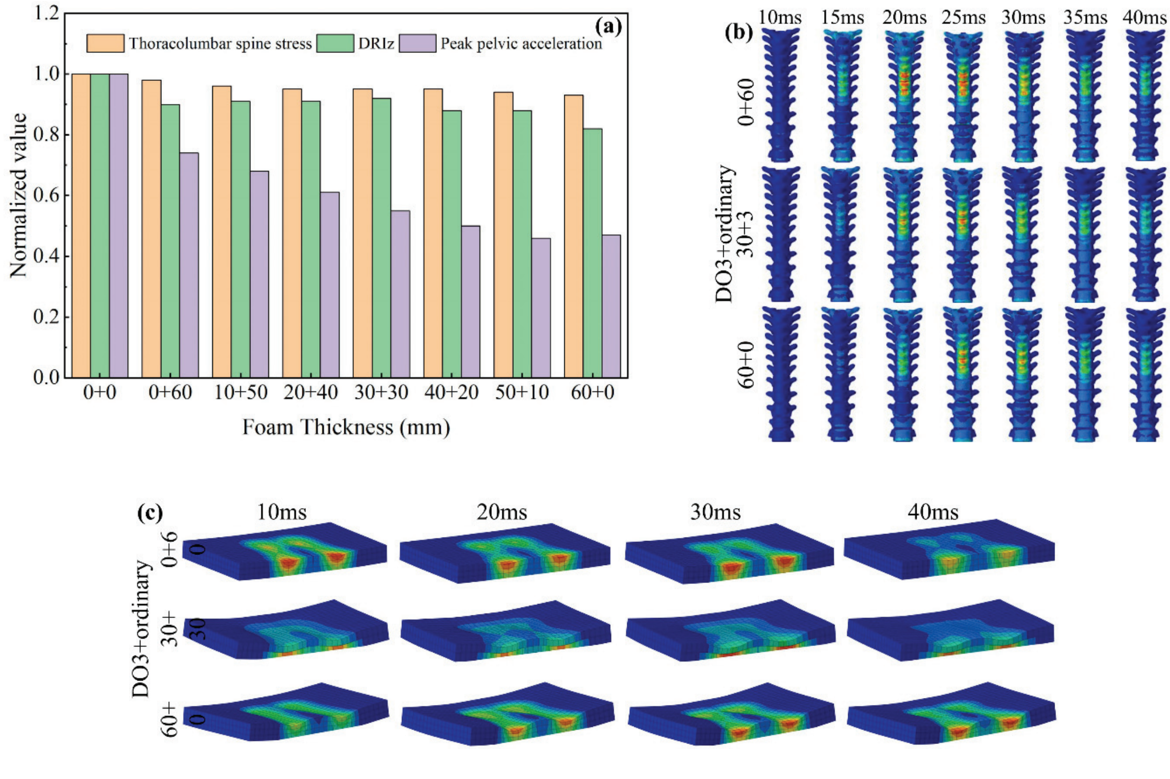

Fig. 7. Normalized maximum lumbar stress, DRIz and peak pelvis acceleration (a), thorax stress distribution (b) and cushion deformation (c) for different cushion designs (the marks such as 30+30 indicate the thickness of DO3 foam+ordinary foam)

with the increase of DO3 foam thickness (Fig. 7b), and the DO3 foam shows more deformation than the ordinary foam (Fig. 7c).

4. Discussion

The human body model validation results (Fig. 3) imply a good bio-fidelity of the C-HBM occupant model for predicting spine response in UBB impacts. The predicted stress concentration behaviour on the thoracolumbar spine (Fig. 6) is in line with the observation from cadaver tests where the spine fractures were mainly occurred in the segments T4–T8 [18]. This also reflects the effectiveness of the C-HUM model in predicting lumbar response. The predicted acceleration curves of the C-HBM occupant model show some deviations out of the PHMS corridors, which might be largely due to the differences in anthropometry, as the C-HBM occupant model represents 50th% Chinese adult male (169 cm in height), which is obviously shorter than the cadavers (177–184 cm), though the potential influence of anthropometry difference on occupant UBB response is not clear, might be induced by the effect of spine length on its stiffness). However, previous studies of human body validation also indicated the difficulty of perfect modelling the PHMS response using FE human body models [7], [19], [24], and cadaver response shows wide diversions in the tests.

The simulation results (Figs. 4–6) of occupant dynamic and biomechanical response in different UBB impacts indicate that reducing the energy of the UBB pulse can significantly lower the risk of occupant thoracolumbar spine and pelvis injury. The detailed biomechanical analysis reveals that when the UBB load input the pelvis of the occupant rotates, which results in stress concentration at the connection between the pelvis and lumbar vertebrae; as the impact energy is continuously transmitted to the spine, anterior bending and axial compression occur at the connection between the thoracic and lumbar vertebrae, and physiological curvature exists at the upper and lower ends of the lumbar vertebrae, which leads to the concentration of stress at the anterior side of the lumbar vertebrae connecting the thoracic vertebrae and the sacral vertebrae (T12–L1 and L4–L5), respectively. The continuous transmission of the UBB load, combined with physiological curve of the human thoracic spine and torso inertia, leads to forward bending and axial compression of the upper thoracic spine, resulting in stress concentration on the T4–T8 segment. These trends are similar to those observed from cadaver tests and FE modelling [4], [18], [19], [25], and the predicted injury mechanism of forward bending combining axial compression is consistent with the conclusions from cadaver tests [18]. It is surprise that spinal peak stress is not sensitive to the change of DRI and maximum pelvic acceleration, which might be largely due to the fact that stress concentration in some elements always exists no matter how the load level was changed. This may suggest that it would be better to use the peak stress together with the stress distribution and peak duration for spine injury risk assessment when applying FE human body model.

Comparisons of cushion protective capability indicate that a cushion can significantly reduce the peak pelvis acceleration but has a less effect in lowing DRIz value and thoracolumbar spine von Mises stress peak, and the DO3 form shows a better protective effect than the ordinary foam (Fig. 7a). These findings could be understood as the following reasons. On one hand, pelvis is the first skeleton part contacting with the seating surface, the acceleration on the pelvis is directly affected by the input load to the human body; the DRIz is not only affected the peak pelvic acceleration but also the compression of the spine (Eqs. (1), (2)), a long duration but low peak acceleration could also lead to a great compression; while the thoracolumbar spine von Mises stress is mainly related to the compression of the spine, the decrease of peak pelvis acceleration from cushion protection has limited effect on reducing the maximum thoracolumbar spine von Mises stress. On the other hand, the DO3 foam made by combining adhesive solution with a polymer can quickly tightens and hardens to digest external forces by immediately lock of the molecules when subjected to severe impact or compression. Thus, the DO3 foam has a better stress–strain properties (Fig. 2c), and shows more deformation than the ordinary foam in the impacts (Fig. 7c), which results in a higher reduction to the pelvis load and hence, a lower pelvis acceleration and DRIz value. The results show that a 60 mm thick DO3 foam could reduce 17.2% of the DRIz value, which may provide a supplementary protective solution for armored vehicle occupant lumbosacral injury in addition to the blast-resistant seats. The above findings may suggest that reducing the pelvic acceleration peak only is not enough for protection of spinal UBB injury risk, but control of torso inertia effect would be much helpful and further necessary, exoskeleton design together with explosion resistant seat may be solution. But this is just the suggested protective measures from the analysis of the current study, and future research on this may provide fundamental guidance for better blast-resistant system design.

It should be noted that several limitations need further improvement or investigation. First, further improvements are still needed to the C-HBM for bettering its biofidelity since the validation results show somehow deviation from cadaver test data. Second, the injury risk of soft tissue is worth analysis with the help of high biofidelity FE human body model. Third, the cushion design only focuses on different materials, detailed structure design could be investigated in future study. Finally, more extensive and in-depth studies could be conducted using the numerical analysis tool and approach to explore more effective protective measures for UBB injury.

5. Conclusions

The current study validated a seated human body model of 50th% Chinese adult male for the application of UBB injury risk prediction, and then the validated human body model was employed for analysis of pelvis and thoracolumbar spine response in UBB impacts and injury protective cushion design. The findings of the current work could be summarized as follows.

1) The C-HUM occupant model has a plausible capability in predicting UBB kinematics and injury risk, which could be applied as a assessment tool for analysis of UBB injury and prevention.
2) UBB impact combined with the effects from physio+logical curve of the human thoracolumbar spine and torso inertia leads to thoracolumbar spine anterior bending and axial compression, which results in stress concentration in the segments of T4–T8, T12–L1, and L4–L5.
3) Foam seat cushion can effectively reduce the risk of thoracolumbar spine injury of armored vehicle occupants in UBB impacts, and the DO3 foam has better protective performance than ordinary foam, the 60 mm thick DO3 foam could reduce pelvic acceleration peak and DRIz value by 52.8 and 17.2%, respectively.
4) UBB spinal injury risk is sensitive to the input load level, but reducing the pelvic acceleration peak only is not enough for protection of spinal UBB injury risk, control of torso inertia effect would be much helpful, which is worth of further research.

Acknowledgements

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References

[1] BAILEY A., CHIRSTOPHER J., BROZOSKI F., SALZAR R., *Post mortem human surrogate injury response of the pelvis and lower extremities to simulated underbody blast*, Ann. Biomed. Eng., 2015, 43 (8), 1907–1917.

[2] BELMONT P., GOODMAN G., ZACCHILLI M., POSNER M., EVANS C., OWENS B., *Incidence and epidemiology of combat injuries sustained during "the surge" portion of operation Iraqi freedom by a U.S. army brigade combat team*, J. Trauma, 2010, 68 (1), 204–210.

[3] COMSTOCK S., PANNELL D., TALBOT M., COMPTON L., WITHERS N., TIEN H., *Spinal injuries after improvised explosive device incidents: implications for tactical combat casualty care*, J. Trauma, 2011, 71 (5, Suppl. 1), S413–17.

[4] DOOLEY C., WESTER B., WING I., VOO L., ARMIGER R., MERKLE A., *Response of the thoracolumbar vertebral bodies to high-rate compressive loading*, Biomed. Sci. Instrum., 2013, 49, 172–179.

[5] HUANG J., HUANG C., MO F., *Analysis of foot-ankle-leg injuries in various under-foot impact loading environments with a human active lower limb model*, J. Biomed. Eng., 2022, 144 (1), 011012.

[6] LI G., MA H., GUAN T., GAO G., *Predicting safer vehicle front-end shapes for pedestrian lower limb protection via a numerical optimization framework*, Int. J. Auto. Tech.-Kor., 2020, 21 (3), 749–756.

[7] LI G., MENG H., LIU J., ZOU D., LI K., *A novel modeling approach for finite element human body models with high computational efficiency and stability: application in pedestrian safety analysis*, Acta Bioeng. Biomech., 2021, 21 (2), 21–30.

[8] LIU X.R., TIAN X.G., LU T., LIANG B., *Sandwich plates with functionally graded metallic foam cores subjected to air blast loading*, Int. J. Mech. Sci., 2014, 84, 61–72.

[9] LSTC. LS-DYNA keyword user's manual, version 971. Liver+ more Software Technology Corporation Livermore, United States of America. 2007.

[10] MA H., MAO Z., LI G., YAN L., MO F., *Could an isolated human body lower limb model predict leg biomechanical response of Chinese pedestrians in vehicle collisions?*, Acta Bioeng. Biomech., 2020, 22 (3), 117–129.

[11] MO F., LI F., BEHR M., XIAO Z., ZHANG G., DU X., *A lower limb-pelvis finite element model with 3D active muscles*, Ann. Biomed. Eng., 2018, 46, 86–96.

[12] MO F., LUO D., TAN Z., SHANG B., ZHOU D., *A human active lower limb model for Chinese pedestrian safety evaluation*, J. Bionic. Eng., 2021, 18 (4), 872–886.

[13] OTT K., DREWRY D., LUONGO M., ANDRIST J., ARMIGER R., TITUS J., DEMETROPOULOS C., *Comparison of human surrogate responses in underbody blast loading conditions*, J. Biomech. Eng., 2020, 142 (9), 091910.

[14] PANDELANI T., CARPANEN D., MASOUROS S., *Evaluating pelvis response during simulated underbody blast loading*, J. Biomech. Eng., 2024, 146 (2), 024501.

[15] POSSELY D., BLAIR J., FREEDMAN B., SCHOENFELD A., LEHMAN R., HSU J., *The effect of vehicle protection on spine injuries in military conflict*, J. Spine, 2012, 12, 843–848.

[16] RUPP J., ZASECK L., MILLER C., BONIFAS A., REED M., SHERMAN D., CAVANAUGH J., OTT K., DEMETROPOULOS C., *Whole body PMHS response in injurious experimental accelerative loading events*, Ann. Biomed. Eng., 2021, 49 (11), 3031–3045.

[17] SCHOENFELD A., GOODMAN G., BELMONT Jr, P., *Characterization of combat-related spinal injuries sustained by a U.S. army brigade combat team during operation Iraqi freedom*, J. Spine, 2012, 12, 771–776.

[18] SOMASUNDARAM K., SHERMAN D., BEGEMAN P., CIARELLI T., MCCARTY S., KOCHKODAN J., DEMETROPOULOS C., CAVANAUGH J., *Mechanisms and timing of injury to the thoracic, lumbar and sacral spine in simulated underbody blast PMHS impact tests*, J. Mech. Behav. Biomed. Mater., 2021, 116, 104271.

[19] SOMASUNDARAM K., ZHANG L., SHERMAN D., BEGEMAN P., LYU D., CAVANAUGH J., *Evaluating thoracolumbar spine response during simulated underbody blast impact using a total human body finite element model*, J. Mech. Behav. Biomed. Mater., 2019, 100, 103398.

[20] STECH E., PAYNE P., *Dynamic models of the human body*, AMRL-TR-66, l–9, 1969.

[21] YOGANANDAN N., HUMM J., BAISDEN J., MOORE J., PINTAR F., WASSICK M., BARNES D., LOFTIS K., *Temporal corridors of forces and moments, and injuries to pelvis-lumbar spine in vertical impact simulating underbody blast*, J. Biomech., 2023, 150, 111490.

[22] YOGANANDAN N., MOORE J., ARUN M., PINTAR F., *Dynamic responses of intact post mortem human surrogates from inferior-to-superior loading at the pelvis*, Stapp Car Crash J., 2014, 58, 123–143.

[23] WEAVER C., MERKLE A., STITZEL J., *Pelvic response of a total human body finite element model during simulated under body blast impacts*, ASCE-ASME J. Risk U. B., 2021, 7 (2), 021004.

[24] WU T., KIM T., BOLLAPRAGADA V., POULARD D., CHEN H., *Evaluation of biofidelity of THUMS pedestrian model under a whole – body impact conditions with a generic sedan buck*, Traffic Inj. Prev., 2017, 18, S148–S154.

[25] ZHANG J., MERKLE A., CARNEAL C., ARMIGER R., ROBERTS J., *Effects of torso-borne mass and loading severity on early response of the lumbar spine under high-rate vertical loading*, Proceedings of the IRCOBI Conference, 2013, 111–123.

[26] ZHANG N., ZHAO J., *Study of compression-related lumbar spine fracture criteria using a full body FE human model*, Proceeding of the 23rd International Technical Conference on the Enhanced Safety of Vehicles (ESV), Seoul, Korea, 2013, Paper No. 13-0288.