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Underbody blast effect on the pelvis and lumbar spine: A computational study

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ABSTRACT

Explosion from an anti-tank landmine under a military vehicle, known as underbody blast (UBB), may cause severe injury or even death for the occupants inside the vehicle. Severity and patterns of lower extremity, pelvis and lumbar spine injuries subjected to UBB have been found highly related to loading conditions, i.e. the vertical acceleration pulse. A computational human model has been developed and successfully simulated the tibia fracture under UBB in the previous study. In the present study, it was further improved by building a detailed lumbar spine and pelvis model with high biofidelity. The newly developed pelvis and lumbar spine were validated against component level test data in the literature. Then, the whole body model was validated with the published cadaver sled test data. Using the validated whole body model, parametric studies were conducted by adjusting the peak acceleration and time duration of pulses produced in the UBB to investigate the effect of waveform on the injury response. The critical values of these two parameters for pelvis and lumbar spine fracture were determined, and the relationship between injury pattern and loading conditions was established.

1. Introduction

An "underbody blast" (UBB) is defined as the detonation of an anti-vehicular (AV) landmine or improvised explosive device (IED) underneath a vehicle. The explosion could result in large elastic and inelastic deformation and acceleration of the vehicle floor to cause injuries/fatalities of the crew. About 67% incidence of injuries involved lower extremity, spine and pelvis during Afghanistan and Iraq wars (Schoenfeld and Dunn et al., 2013). Compared to the studies on the lower limb injuries, the work on pelvis and lumbar spine is rare, due to the structural complexity of these body regions, and the complicated boundary and loading conditions. Characterization of how the acceleration pulse associated with an UBB affects the response of lumbar spine and pelvis will provide the valuable information towards understanding the relevant injury mechanisms and threshold. With the information, protective devices for mitigating AV landmine and IED effect can be further developed.

The biomechanical response of pelvis and lumbar spine under UBB can be studied through physical tests using either ATD (Anthropomorphic test device, also known as crash test dummy) (Baudrit et al., 2005; Polanco and Littell, 2011) or post mortem human subjects (PMHS) (Bailey et al., 2013; Bailey et al., 2015; Lehman et al., 2012; Stemper et al., 2011; Spurrier et al., 2015; Yoganandan et al., 2015). In these tests, the acceleration pulses generated by UBB were usually simulated by a high-speed impact load applied to the seating subjects in the vertical direction. Then the biomechanical responses of the relevant body regions can be measured. The surrogates used in the tests, however, have some limitations. For example, ATDs are made from artificial materials without anatomical details. Due to its limited biofidelity, dummies are not able to predict bone fractures. On the other hand, PMHS can produce tissue level injuries. But due to its high cost, large data scatter and difficulty in instrumentation, tests with human cadavers are not extensively conducted.

As a valuable supplement to the physical tests, numerical simulations (frequently with finite element, or FE method) can overcome the limitations of physical tests. A sufficiently validated human body model may have good biofedility and produce highly reliable data at low cost. More importantly, it can predict the responses which are not possible to obtain experimentally, such as stress, strain of soft tissue and energy transfer. A number of computational studies were conducted on the UBB effect of human body. For example, Zhang et al. (2013) built a partial human body FE model with lumbar spine, pelvis and femur to simulate the response under high-rate vertical loading. The results show that the high-speed vertical loading in the spine produces an 'S' shaped

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deformation, indicating the change of injury mechanisms along the spine. Instead of using a partial body model, Zhang and Zhao (2013) developed a detailed lumbar spine FE model and integrated it into a 50th percentile full human body model. Vertical loading was applied to the bottom of the human body. The injury patterns and injury criteria for the lumbar vertebral fractures under compressive-dominated loadings were investigated. However, the injury response of pelvis was not studied.

Based on the previous studies, one can see that the numerical simulations of UBB effect on the human body are still very much limited, and no dedicated full body models under UBB loading are available. To resolve this issue, in the current study, we improved an existing full body model developed for blast-induced lower leg injury study and extended it to investigate pelvic and lumbar spine injuries. A new lumbar spine and pelvis model was built and integrated into the torso. Then component and whole body level validations were performed on the newly developed pelvis/lumbar spine models and the full body model, respectively. After that, a comprehensive parametric study was performed to establish the relationship of injury patterns/threshold and impact parameters.

2. The finite element model

2.1. Mesh development

A finite element human model was developed and successfully simulated the tibia fractures under UBB. (Dong et al., 2013). The model had a highly biofidelic lower extremity, and a simplified pelvis and spine was used to reduce the computational cost. In the present study, it was further improved by including a detailed lumbar spine and pelvis model with anatomic details. The pelvis and lumbar spine were developed based on the computed tomography (CT) images of a 50th percentile male (Gayzik et al.,2011). The bony structures and soft tissues were mainly modeled with high quality hexahedral elements. The nodes at the interfaces between different body regions were merged and thus the force transfer can be calculated more accurately.

The average mesh size of pelvis and lumbar spine was 3 mm. The total numbers of solid and shell elements of the whole body model were 913,924 and 210,713 respectively. The whole body model was integrated with a highly simplified seat and foot plate and set to have a normal posture of occupant. The setup of whole body model is illustrated in Fig. 1a. The enlarged views of the original and improved lumbar spine and pelvis are shown in Fig. 1b and c, respectively. The new model was built with the meshing software Hypermesh (Altair, Troy, MI) and all simulations were performed using LS-DYNA971 (LSTC, Livermore, CA).

2.2. Material properties

Constitutive laws and corresponding material constants for all body parts were carefully selected according to the biological tissue's material behavior. Under the loading of UBB, due to the large acceleration in a very short time duration, the bony structures are expected to undergo high strain rates, which can further cause fracture. Therefore, the material laws to describe the behavior of bones should include the strain rate effect and failure criterion. In this model, the bony materials were described using an elastic-plastic constitutive law, i.e. Mat 81 (MAT_PLASTICITY_WITH_DAMAGE) in LS-DYNA (LS-DYNA keyword User's manual). Since the strain rate related parameters for some bones including pelvis and lumbar spine are not available in the literature, the material constants for femur were used for all of the cortical bones. (Dong et al., 2013) The failure criterion for bones was assumed to be based on strain. When the maximum plastic strain exceeds the failure threshold, the elements involved are considered 'failed' and deleted from the model, to simulate the bone fracture. This approach has been widely used in the modeling of traffic and military injuries (Song et al., 2005; Zhang and Zhao, 2013). In the present study, the fracture of trabecular bones was neglected, since cortical bones take most of the loads (Song et al., 2005).

The soft tissues, such as flesh and skin were modeled as visco-elastic materials, i.e. MAT_92 (MAT_SOFT_TISSUE_VISCO) (LS-DYNA keyword User's manual). The input parameters of major materials in the pelvis and lumbar spine are listed in Table 1.

3. Numerical model validation

The new model was validated at both component (body part) and whole body levels. At the component level, impact tests on the isolated pelvis (Guillemot et al., 1997; Song et al., 2005) and lumbar spine (Stemper et al., 2011) available in literature were simulated. Then a published sled test on cadavers (Yoganandan et al., 2014) was used to validate the whole body model.

3.1. Pelvis validation

Guillemot et al. (1997) studied the damage behavior of 12 cadaveric pelvic bones under side impacts. A metallic ball of 3.68 kg impacted the pelvic acetabulum at a speed of 4 m/s in all of the tests. The new pelvis model was used to simulate these tests, and the model setup is shown in Fig. 2a. The loading and boundary conditions were identical to those in the physical tests. In the both experiments and simulations, the time histories of contact force between the steel ball and acetabular and displacement of the ball were monitored. Then the force-displacement relationship was obtained by combining these two curves. The

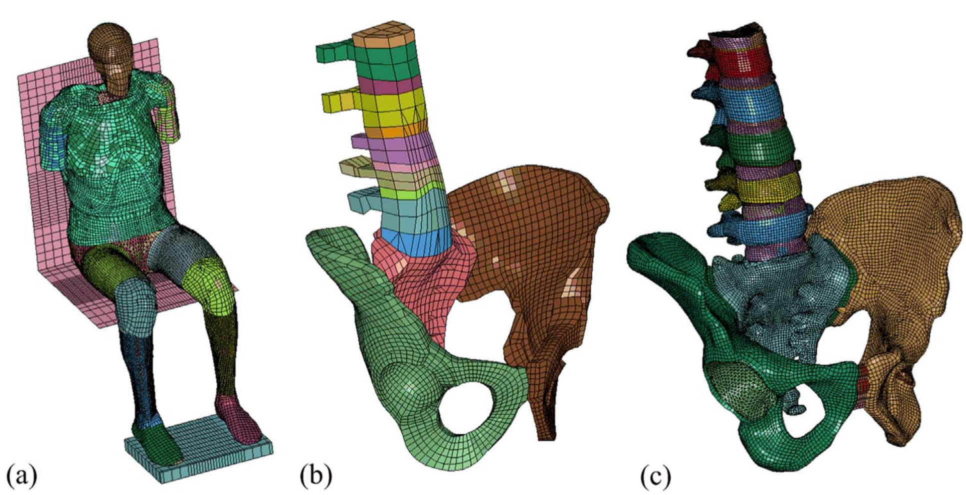

Fig. 1. (a), Whole body model used for UBB loading simulations. (b) Enlarged view of the original lumbar spine and pelvis models in Dong et al. (2013). (c) Enlarged view of improved lumbar spine and pelvis models developed in the current study.