Pelvic Response of a Total Human Body Finite Element Model During Simulated Under Body Blast Impacts

Caitlin M. Weaver and Joel D. Stitzel

Abstract Under body Blast (UBB) events seen in theater are the cause of many serious injuries sustained by soldiers in combat zones to the pelvis, spine, and lower extremities. These injuries are often debilitating, resulting in increased healthcare expenses and a reduced quality of life. Injury prediction for UBB events continues to be a challenge due to the limited availability of UBB-specific test studies and injury criteria.

This study focuses on the pelvic injury response of the 50th percentile male Global Human Body Models Consortium (GHBMC) Finite Element (FE) model. Evaluation of the GHBMC model fidelity and injury response is based on biofidelity targets (corridors) created using pelvis accelerations obtained from experimental testing of UBB-type loading using post mortem human subjects (PMHS). The FE acceleration data extracted from nodes in the S1 region of the GHBMC pelvis was compared to the S1 experimental biofidelity corridors created from this experimental PMHS test data. The FE S1 acceleration was analyzed with an objective rating method (CORrelation and Analysis, CORA) using these experimental biofidelity corridor curves. The CORA analysis showed good correlation (0.70 or higher) with average scores ranging from 0.841 to 0.808 using FE and experimental data filtered at 1050Hz.

Keywords finite element model, GHBMC, pelvic injury, under body blast, WIAMan

I. INTRODUCTION

Military operations in Iraq and Afghanistan over the past several years have resulted in the increased exposure of military personnel to improvised explosive devices (IEDs) and road side bombs. Blast injury refers to all injuries that can occur as a result of exposure to an explosive mechanism [1]. Under-body blast (UBB) is a type of blast event in which exposure to an explosive mechanism, typically in the form of IEDs or road side bombs, occurs to occupants of combat vehicles. UBB events are also characterized by high rate, short duration vertical loading, resulting in a wide range of physical injuries for which causation and mitigation are not fully known or understood. Injury to the occupant occurs as a result of the transfer of blast load energy from the vehicle floor and seat to the occupant [2-3].

The use of explosive mechanisms as combat weapons has resulted in a significant number of blast-related casualties. Since WWI there has been an increasing trend in injury by explosive mechanism [1]. Injury data studies have reported an increase in injury due to explosive mechanism from 35% in WWI to 79% in Operation Iraqi Freedom (OIF) and Operation Enduring Freedom (OEF). This injury data study also showed that 38%, or nearly half, of all explosive-related casualties in OIF/OEF were the result of an IED. [1][4-6].

Current military Live-Fire Test and Evaluation (LFT&E) efforts for armored vehicles safety during UBB events use anthropomorphic test devices (ATDs) that are primarily designed and validated to predict occupant safety for civilian motor vehicle accidents. As a result, injury prediction for UBB events continues to be a challenge due to the limited availability of accurate UBB-specific test studies. UBB injury prediction methods are subject to injury criteria developed for motor vehicle crash injury prediction, often limited to automotive loading rates. In an effort to improve UBB LFT&E injury prediction, the Department of Defense (DoD) approved the Warrior Injury Assessment Manikin (WIAMan) Project [2].

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The purpose of the WIAMan effort is to create an enhanced capability to assess risk to soldiers in the UBB environment for use in LFT&E and protection technology development. This includes the creation of a soldier-representative, biomechanically-validated anthropomorphic test device (ATD) [7]. Injury assessment reference values (IARVs) for this ATD will be developed for core body regions, including the pelvis, which were selected as a result of an injury data analysis performed by the Joint Trauma Analysis and Prevention of Injury in Combat (JTAPIC) partnership. This study was performed using a data set of 608 wounded in action (WIA) and killed in action (KIA) casualties with 1,637 and 2,912 injuries, respectively, coded using the Abbreviated Injury Scale (AIS) [2][7-8].

The data presented from this JTAPIC study reported 32% of KIA casualties and 18% of wounded in action WIA casualties suffered tibia and fibula injuries. This study also showed a high occurrence of foot and ankle injuries with 32% of KIA casualties and 26% of WIA casualties suffering from injury to this region. Though pelvic fractures were only present in 5% of WIA casualties, 46% of KIA casualties suffered from pelvic fracture. The high mortality associated with pelvic fractures could be the result of (1) comorbidities resulting from the severity of the blast event or (2) injury due to inability to perform load bearing activities necessary to safely evacuate a vehicle after a blast event to avoid further threat and to receive treatment [2][9]. Literature data from casualty studies along with preliminary UBB-simulated PMHS sled tests report blast induced pelvic fractures in the pubic rami, ischium, sacral altar, acetabulum, and sacroiliac regions of the pelvis [9-11]. With the exception of pubic rami fractures, all of these fractures are classified as partially stable or unstable based on the coding system developed by the Orthopedic Trauma Association (OTA) [12].

Pelvic injuries both in the civilian and military environment are often debilitating, resulting in increased healthcare expenses and a reduced quality of life. According to a ten year injury data study conducted on Crash Injury Research and Engineering Network (CIREN) data from civilian motor vehicle accidents, median in-hospital medical charges were higher for patients with injuries to the pelvis and lower extremities than for patients without these injuries. This study also reported that pelvic fracture had the largest total median in-hospital charge cost, with acetabulum fracture resulting in the highest median in-hospital charge cost ($20,723) of all fractures occurring in the pelvis and lower extremity region. The high direct economic costs associated with these injuries do not include any indirect costs, such as the inability or delay in return to pre-injury activities [13-14]. Pelvic fractures in a combat setting also pose critical issues related to trauma care. Upon injury, early pelvic stabilization is necessary to control hemorrhage and reduce mortality. Partially stable and unstable pelvic fractures pose a challenge for quick and safe vehicle evacuation, as well as combat casualty care (CCC) [15].

The majority of reported lower extremity injury evaluation studies for UBB-type loading only involve biomechanical testing on the lower extremity region below the knee. The first published attempt to investigate pelvis injury response in PMHS using UBB-type loading was performed on whole body PMHSs using a dual, independent sled blast simulator [9]. Other UBB injury focused studies have been conducted using both PMHS whole body [16] and component experimental testing [17] and FE simulation [3][16][18]. These studies, however, have only focused on the lumbar and cervical regions of the spine. Additionally these studies were only performed on FE body components and not on a full human body FE model.

This study focuses on the pelvic injury response of the 50th percentile male Global Human Body Models Consortium (GHBMC) FE human body model. The GHBMC was chosen for the human body modeling portion of this study because it is a high-fidelity model that was developed to investigate human body response during dynamic impact events and has been validated in various impact scenarios [19]. The GHBMC model has been used in several studies to investigate human body response during dynamic impact events. To date, the comparison of full body UBB experimental testing to drive and compare with full body FE simulation metrics for UBB is unique. This study is a preliminary attempt to validate the GHBMC for pelvic injury resulting from UBB-type loading using the metric of S1 acceleration produced in PHMS experimental testing and FE simulations. This data was acquired with the explicit purpose of developing an enhanced capability to predict the risk of injury for mounted soldiers who are subjected to the effects of UBB loading with the goal of enhanced vehicle and Soldier survivability.

II. METHODS

This study was performed using FE simulations in LS-DYNA software and input data from vertically accelerative load testing performed on PMHS by the Biomechanics Product Team (BIO PT) for the U.S. Army WIAMan project. The S1 outputs of these simulations were analyzed using an objective rating method (CORrelation and Analysis, CORA) performed using preliminary biofidelity corridors.

The FE rig test configuration for this study was developed using LS-DYNA (version 6.1.1, revision 78769) and is based on a simplified model of the test configuration for UBB designed by Johns Hopkins University Applied Physics Laboratory (JHU-APL) using the Vertically Accelerated Load Transfer System (VALTS) rig (Fig. 1). VALTS was designed to model the UBB environment in a laboratory test setting. The design of this system allows for the independent application of controlled pulse duration to the seat and the floor of the system. The system also contains mounted cameras for kinematic tracking. Restrain systems are added to the VALTS seat. For these tests a five-point harness was used to restrain the test specimens.

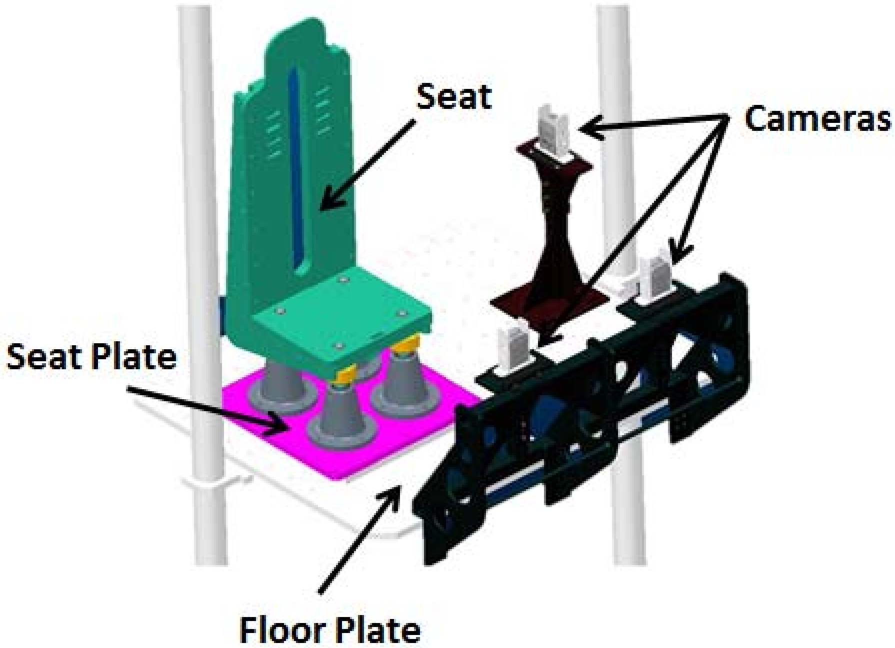

Fig. 1. VALTS.

The FE test rig used was a simplified design of the VALTS rig consisting of three components: (1) a seat with a seat back and head rest, (2) a floor plate, and (3) a five-point harness (Fig. 2 and Fig. 3). The GHBMC human body FE model was used to represent the PHMS specimens used in experimental testing. The GHBMC is licensed and distributed for academic and commercial uses by Elemance, LLC. The computing system used to run the simulations for this study was the Distribution Environment for Academic Computing (DEAC) cluster, a Linux Red Hat 6 high performance computing system.

The acceleration pulse curves provided by the WIAMan BIO PT were used to move the seat and the floor plate along the Z-direction using *BOUNDARY_PRESCRIBED_MOTION_RIGID (Fig. 2 and Fig. 3). Nodes selected on the bottom portion of the seat and the floor plate were used to extract data using *NODOUT. The FE *NODOUT velocity data was compared to the experimentally recorded velocity data to determine if the FE rig had a similar response to the experimental rig.

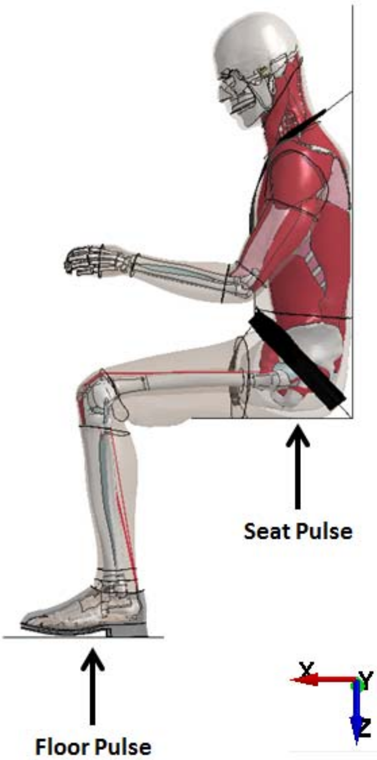

Fig. 2. Test rig designed with belted GHBMC (side view).

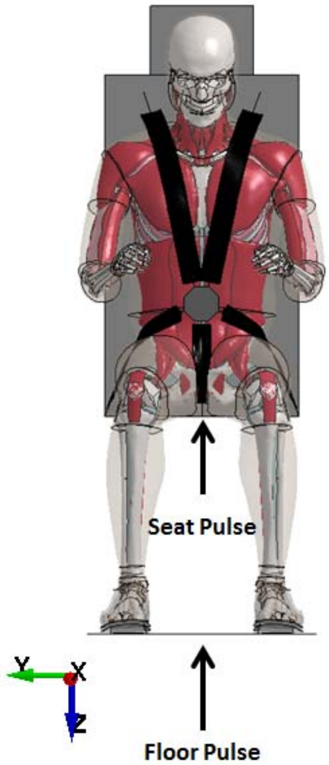

Fig. 3. Test rig designed with belted GHBMC (front view).

The default position for the current version of the GHBMC (v4.3) is for FE simulations of a seated driver in a civilian motor vehicle [19]. To change the model's position to fall within the pre-test positioning guidelines of the WIAMan BIO PT, a joint repositioning method was employed in LS-DYNA to adjust the knee from the initial angle of approximately 120^{o} (Fig. 4) to approximately 90^{o} (Fig. 5). The repositioning method used *BOUNDARY_PRESCRIBED_MOTION_NODE to translate the legs in the x, y, and z-direction to achieve the desired knee angle and heel-to-heel distance (Fig. 6). These positions were measured in LS-PREPOST. Due to issues with run time and element distortion, a simplified model of the legs with rigid material used for the interior lower limb and knee flesh was used to adjust the knee angle. The lower limb model was reincorporated into the full body model.

Fig. 4. Leg model at original 120^{o}.

Fig. 5. Leg model repositioned to 100^{o}.

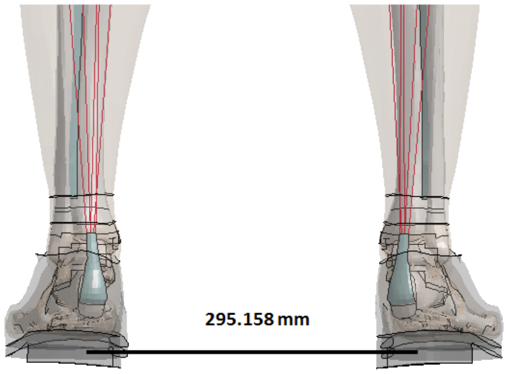

Fig. 6. Heel to heel distance for repositioned legs.

The pelvis angle of the GHBMC also needed to be adjusted to fall within the pre-test positioning guidelines. For this testing, the pelvis angle is determined by the position of landmarks on the pelvis, specifically, the anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), and pubic symphysis (PS) (Fig. 7). The pelvis angle used for positioning is the angle from vertical to the line created by the ASIS and pubic symphysis. Based on nodal position obtained in LS-PREPOST, the initial angle of the GHBMC pelvis was 21.571°. According to the pre-test positioning guidelines, the pelvis angle must be between 35° and 45°. A repositioning method was performed in LS-DYNA to adjust the angle of the pelvis to approximately 36°. The repositioning method used *BOUNDARY_PRESCRIBED_MOTION_SET to rotate the pelvis about the y-axis to achieve the desired pelvis angle. The x-z coordinates used for the rotation were the coordinates for the center of mass of the pelvis. Fig. 8 and Fig. 9 show the initial and adjusted pelvis angle of the GHBMC.

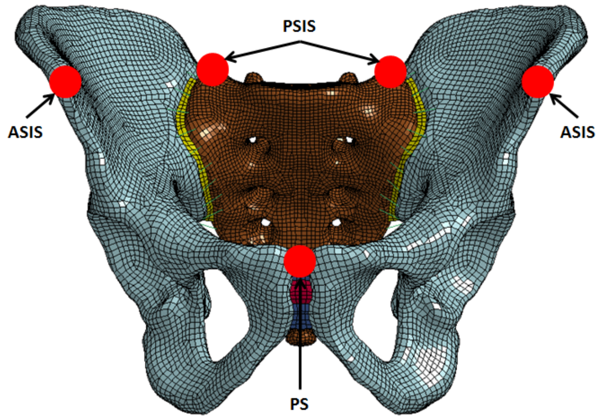

Fig. 7. Pelvis landmarks used for positioning.

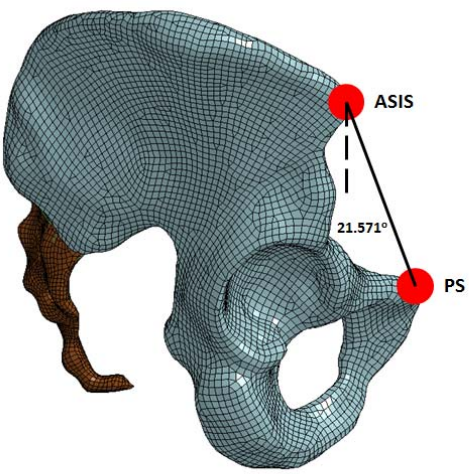

Fig. 8. GHBMC pelvis at original position.

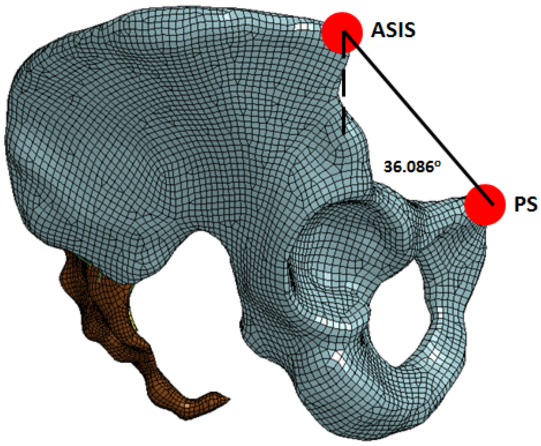

Fig. 9. GHBMC pelvis at adjusted position.

The input data used for this FE study was obtained from vertically accelerative load testing performed on PMHS by the WIAMan BIO PT. The tests used in this study applied a 4m/s pulse to the seat and floor plate. The UBB FE simulations were conducted by applying acceleration pulses to the floor and seat of the FE rig. These pulses were obtained from accelerometers attached to the floor and seat of experimental test vehicle rigs. Fig. 10 and Fig. 11 show an example of one of the test curves used for this study for seat and floor acceleration, respectively. Though these curves are not the exact curves used in the simulation, they show the peak acceleration and time to peak for this test. For these simulations, the GHBMC was positioned in a FE vehicle rig seat in a similar configuration to the PMHS used for experimental testing. The position of the heels, pelvis angle, vertical distance between the posterior aspect of the C7 vertebrae and the average location of the ASIS, and the angle between the lap belt and seat was within the measured tolerances used to position the PMHS. The GHBMC was gravity settled and the belts were tightened with an 180N pretension load for 110ms. The acceleration pulses were applied to the floor and the seat for 40ms. The time step used for these simulations was 3.00E-04ms.

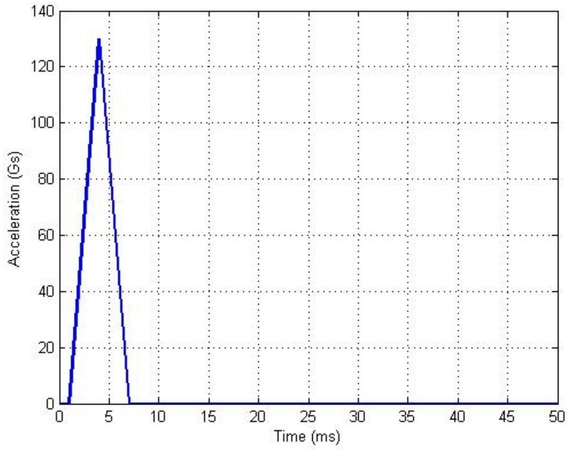

Fig. 10. Example seat pulse.

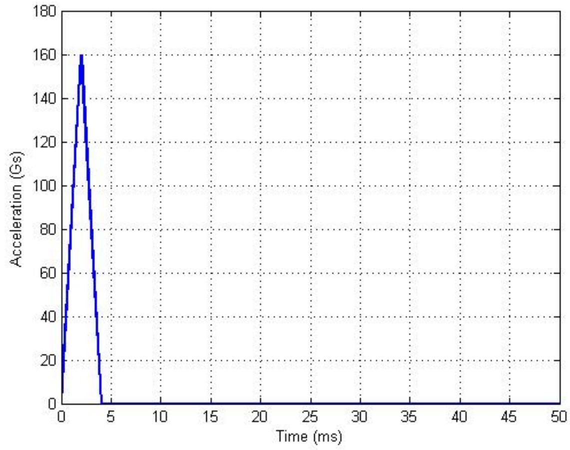

Fig.11. Example floor pulse.

Acceleration data from 133 nodes in the S1 region of the pelvis of the GHBMC were extracted from the simulations. These nodes were chosen because they were in the S1 region of the pelvis and were the same surface area as the DTS 6DX PRO accelerometer (Diversified Technical Systems, Inc.) used in PMHS experimental testing. The sampling rate for the experimental testing ranged from 100 kHz to 1 MHz, depending on the test. Fig. 12 shows the coordinate system for pelvis orientation for this testing. The extracted FE S1 acceleration data was compared to S1 data recorded from the experimental biofidelity corridors created from preliminary WIAMan experimental test data. The direction used for this comparison was SAE-Z. Corridors were created using data filtered 1050Hz. The corridors were generated using a standard approach determined by a Biofidelity Response Corridor (BRC) working group. The approach aligns non-normalized signals using the Nusholtz method, which transforms signals to principal component space using eigenvectors and eigenvalues, and generates ±1 and ±2 standard deviation equivalent corridors. The corridors used for this study were constructed using data from 13 tests [20].

The frequency value used to create this corridor was determined from preliminary work performed by the Signal Analysis Working Group (SAWG) for the WIAMan project. The SAWG is investigating the adequate range of optimal filter frequencies for analysis of biomechanical signals. The method for filtering employed by the SAWG is a four pole, zero-phase Butterworth filter. The purpose of this filtering technique is to determine an ideal filter frequency for individual signals in a set in order to customize the filter for varying test configurations and speeds. This method determines the ideal frequency of each signal using the peak impact magnitude. The signal frequencies are then compiled and a mean frequency of all the signals is determined. For this PMHS test data, 1050Hz is the average frequency at which the change in frequency with respect to peak magnitude begins to level off for each data trace. Evaluation of GHBMC model fidelity and injury response is based on biofidelity targets (corridors) created using pelvis accelerations obtained from experimental testing of UBB-type loading using post mortem human subjects (PMHS).

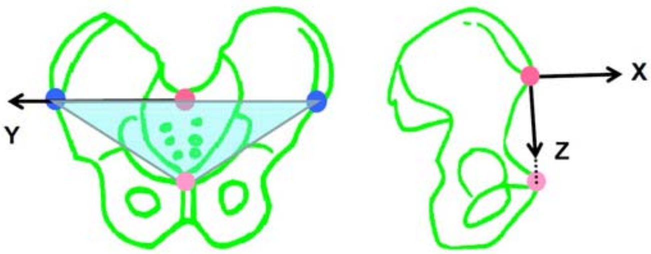

Fig. 12. S1 accelerometer orientation and coordinate system.

III. RESULTS

The FE S1 acceleration showed good correlation with the preliminary biofidelity corridors that were created using the ±1 and ±2 standard deviation equivalent corridors generated using BRC working group method. An analysis was performed using an objective rating method (CORrelation and Analysis, CORA) using the preliminary biofidelity corridors generated corridor curves on four FE tests. These tests were performed with seat and floor pan velocities of 4m/s. Acceleration data for these tests were recorded at different sampling frequencies ranging from 100 kHz to 1 MHz.

CORA evaluates the level of correlation between test and simulation results. This method combines two independent sub-methods: (1) a corridor rating and (2) a cross-correlation rating (Fig. 13). The rating results for both sub-methods range from 0 (no correlation) to 1 (perfect correlation). The corridor rating method calculates the deviation between the curves by evaluating the curve fitting of a response curve into corridors. These corridors can be user-defined or automatically generated. The deviation is calculated using corridor fitting. To perform this corridor fitting, a mean curve is calculated and two corridors, the inner and outer corridor, are defined along the mean curve. If the evaluated curve fits within the inner corridor, a score of 1 is given. The score decreases from 1 to 0 between the bounds of the inner and outer corridor. This deviation is calculated for each time step with the final rating of the signal calculated as the average of all the time step ratings. The cross-correlation method evaluates the characteristics of the signal. This method analyzes the differences between the curves using three sub-methods – phase shift, size, and shape – with individual ratings. The scores from the corridor rating and the cross-correlation rating are combined to determine the final CORA score [21].

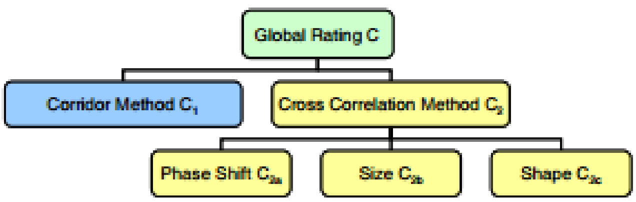

Fig. 13. Structure of the CORA rating scheme.

The level of correlation between the FE test data and the preliminary BRCs were analyzed using CORA. The corridors were constructed using user defined inputs of the preliminary corridors generated using PMHS experimental tests. The ±1 and ±2 SD curves were used for the inner and outer corridor limits, respectively. The average corridor curve was used as the cross-correlation reference. The CORA analysis showed good correlation

(0.70 or higher) with an average of 0.841 with a maximum time interval of 10ms, 0.844 with a maximum time interval of 15ms, and 0.808 with a maximum time interval of 20ms. The results of the CORA analyses for these FE simulations with a maximum time interval of 15ms are shown in Figs. 14-17. For these figures, the cross-correlation reference curve is represented in yellow, the inner corridor limit curves are represented in green, the outer corridor limit curves are represented in blue, and the FE simulation data curve is represented in red. Table 1 shows the CORA scores produced by these test for this CORA rating method.

TABLE 1
CORA SCORES FOR TEST RESULTS

Test Number	CORA Score (10ms)	CORA Score (15ms)	CORA Score (20ms)
1	0.876	0.865	0.837
2	0.774	0.792	0.768
3	0.831	0.831	0.768
4	0.883	0.889	0.858
Average	0.841	0.844	0.808

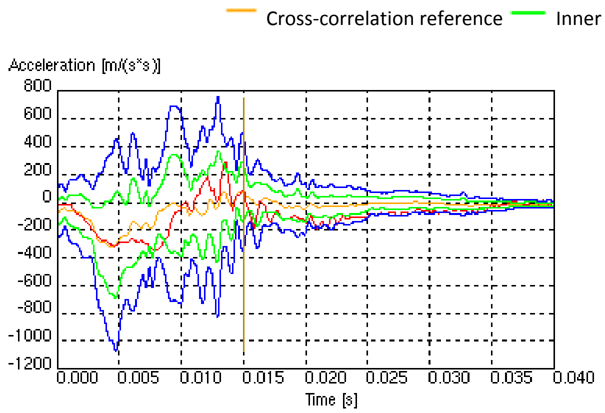

Fig. 14. Test 1 CORA curve comparison with a maximum time interval of 15ms with a score of 0.865.

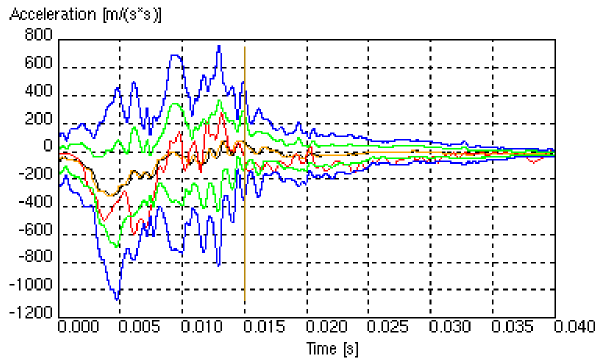

Fig. 15. Test 2 CORA curve comparison with a maximum time interval of 15ms with a score of 0.792.

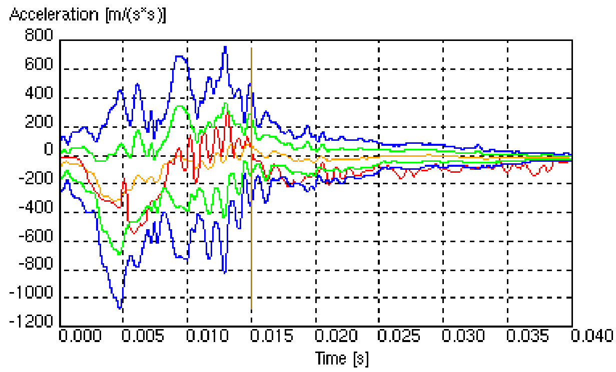

Fig. 16. Test 3 CORA curve comparison with a maximum time interval of 15ms with a score of 0.831.

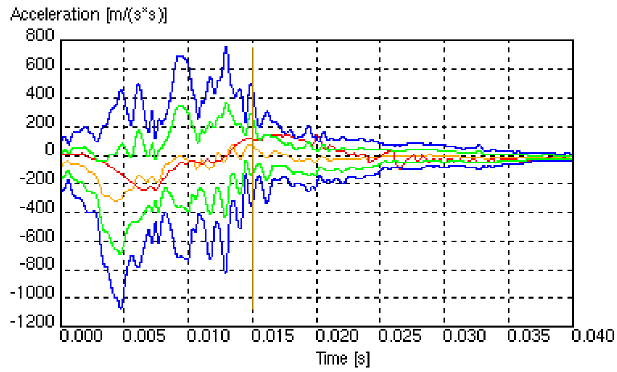

Fig. 17. Test 4 CORA curve comparison with a maximum time interval of 15ms with a score of 0.889.

IV. DISCUSSION

Injuries to the pelvis, spine, and lower extremities are frequently sustained as a result of UBB events. Though several studies have been conducted in an attempt to investigate the effects of UBB loading on the lower extremities, there has not been a significant amount of research studies involving these effects on the pelvis. The current focus of biomechanical research on the pelvis has been on automotive rate loading. This focus presents an issue for UBB related injuries because common injuries such as pelvic ring and ischium fractures are not commonly caused by automotive rate loading. As a result, it is important to conduct research studies on the pelvis involving UBB type loading to understand the causation and mechanisms of these types of pelvis injuries [9].

The initial results for this FE study have shown good correlation for results comparison between PHMS experimental testing and FE human body model acceleration outputs of the S1 region of the pelvis. There are differences in correlation values between the tests analyzed in these studies using the preliminary biofidelity corridors created using the WIAMan BIO PT data. The differences in these results are most likely due to the construction of the preliminary biofidelity corridors. These corridors were constructed using tests conducted on PHMS using the same seat and floor velocity pulses, but with various levels of personal protective equipment (PPE). Since certain PPE, such as body armor, increases body-borne mass, it could have an effect on the forces experienced by the occupant in an UBB event. To quantify this effect, future work should include CORA analyzes on tests using biofidelity corridors constructed using the same levels of PPE. The pelvic response in the FE model shows translation in the superior-inferior (Z) direction as well as the anterior-posterior (X) direction. The translation observed in the Z direction is expected because the acceleration pulses are applied to the seat and floor of the rig in this direction. Translation observed in the X direction is greater at the iliac wings than in other regions, namely the S1 and pubic symphysis regions. These observations are preliminary and need to be further compared to post-positioning measurements from the experimental PMHS testing.

Since this study analyzes only one injury metric using four test and preliminary BRCs, the results from this study should be considered preliminary. Further simulations need to be performed evaluating the pelvis using additional tests and finalized BRCs. Additional injury metrics as well as further characterization using additional parameters, such as stress, strain, and force will need to be analyzed before full validation of the GHBMC for UBB can be determined.

V. CONCLUSIONS

This study focuses on the pelvic injury response of the 50th percentile male Global Human Body Models Consortium (GHBMC) FE human body model. This study is a preliminary validation of the GHBMC for pelvic injury resulting from UBB-type loading using the metric of S1 acceleration produced in PHMS experimental testing and FE simulations. This study was performed using FE simulations in LS-DYNA software and input data from vertically accelerative load testing performed on PMHS by the Biomechanics Product Team (BIO PT) for the U.S. Army WIAMan project. The initial results for this FE study have shown good correlation for results comparison between PHMS experimental testing and FE human body model acceleration outputs of the S1 region of the pelvis. Further FE UBB impact simulations and additional human body model metrics will be compared to the experimental biofidelity corridors. To date, the comparison of full body UBB experimental testing to drive and compare with full body FE simulation metrics for UBB is unique.

VI. ACKNOWLEDGEMENT

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