Evaluation of Biofidelity of Finite Element 50th Percentile Male Human Body Model (GHBMC) under Lateral Shoulder Impact Conditions

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Abstract The goal of this study was to evaluate the biofidelity of a finite element 50th percentile male human body model (GHBMC) under lateral shoulder impact loading conditions by comparing its responses to those of post mortem human surrogates (PMHS) from literature. The GHBMC model (version: FMB v.4.1.1) was positioned on a rigid seat and the right shoulder of the model was impacted by a rigid impactor in different initial impact speeds and directions (impact speeds: 1.5, 3.0, and 4.4m/s, impact direction: 0^{o}, -15^{o}, and $+15^{o}$). The GHBMC model showed similar impact force time histories to those of the PMHS for the various loading conditions, but predicted less peak shoulder deformation and energy dissipation than those of the PMHS. Since injury risk functions for the shoulder use the shoulder deflection as an injury predictor, the GHBMC model needs to be improved to be able to predict the shoulder injury of PMHS. Therefore, further effort is required to validate the shoulder region of the GHBMC model with focusing on its deformation.

Keywords GHBMC, human body model, biofidelity, shoulder, lateral impact

I. INTRODUCTION

In an effort to develop a biofidelic computational surrogate, the finite element (FE) 50th percentile male GHBMC human body model has been developed by Global Human Body Models Consortium™ and validated in various loading conditions [1-7]. Park et al. [8] evaluated the biofidelity of the GHBMC model under the lateral rigid wall impact condition by comparing its response to that of Post Mortem Human Surrogates (PMHS). The model showed good biofidelity in terms of external force while the shoulder kinematics, lower extremity response and fracture prediction showed the need of improvement. Moreover, the shoulder of the GHBMC model sustained far less deformation (relative distance from T1 to acromion) than that of the PMHS. Park et al. claimed that the whole body kinematics of the model was likely to be influenced by the response of the shoulder to the lateral impact, since the shoulder and the pelvis were the primary load paths under their lateral impact condition; they highlighted the need for further validation work focused on the shoulder response of the model. To our knowledge, there are no extant studies on the shoulder biofidelity of the GHBMC model under the lateral impact condition.

Several researchers have conducted lateral and oblique shoulder impact tests using PMHS. Irwin et al. [9] and Koh et al. [10] characterized the response of the shoulder in sled tests; Koh et al. additionally developed an injury risk function for the shoulder using the deflection of the shoulder (T1 to shoulder edge) as the injury predictor. Bolte et al. [11] conducted PMHS shoulder impact tests using a pneumatic impacting ram at the level of the glenohumeral joint with an impact speed at approximately 4.4 m/s for lateral and oblique impact directions (-15^{o}, $+15^{o}$) and developed a force-deflection corridor for the shoulder. Compigne et al. [12] performed PMHS shoulder impact tests at both non-injurious (1.5 m/s) and injurious (3 to 6 m/s) conditions for three different impact directions: -15^{o}, 0^{o}, $+15^{o}$ (Figure 1). They proposed a force-deflection corridor of the shoulder and also analyzed the kinematics of the shoulder complex. Subit et al. [13] impacted the shoulder of the PMHS at constant speed (1 m/s, 3 m/s) in the same impact directions as in Compigne's tests and characterized the force-deflection curve of the shoulder. During that test, the detailed kinematics of the scapula, sternum and two thoracic vertebrae (T1 and T8) were recorded using the VICON™ motion capture camera system.

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The goal of this study was to evaluate the biofidelity of the GHBMC shoulder for various loading conditions in terms of impact velocity as well as the impact direction by comparing its response to that of PMHS from the tests conducted by other researchers.

II. METHODS

In order to compare the shoulder response of the GHBMC model to that of PMHS, three different PMHS tests conducted by Bolte et al. [11], Compigne et al. [12] and Subit et al. [13] were introduced as the reference response of the model to the shoulder side impact condition. The test conditions used in those studies are summarized in Table I, while more detailed information can be found in the references [11-13]. The impact force and shoulder deflection response of the GHBMC model were compared to those of PMHS from the tests. Moreover, the kinematics of the shoulder with respect to the seat and thoracic spine (T8) from Subit's test was analyzed and the shoulder kinematics of the GHBMC model in that test condition was compared to that of PMHS. LS-DYNA double precision MPP R6.1.1 was used as the FE solver of the simulation.

TABLE I
PMHS shoulder impact tests introduced in this study

Test conducted by	Impact velocity [mm/s]	Impact direction [^{o}]	Number of tests (Male/Female)	Impact location	Impactor Shape (W*H) [mm] / mass [kg]	Remarks
Bolte et al. [11]	4.4	0 (lateral)	4 (3/1)	level of glenohumeral joint	200*150 / 23	Impactor head is covered by Arcel 310
Compigne et al. [12]	1.5	0 (lateral)	7 (2/5)		150*80 / 23.4	
		-15 (oblique)				
		+15 (oblique)				
Subit et al. [13]	3.0	0 (lateral)	2 (2/0)	-24mm (measured)	400*75 / 71.8	※ Impact stroke : 75[mm] ※ Impact location: the measured impact position of the impactor relative to the level of glenohumeral joint in vertical direction
		-15 (oblique)		-64mm (measured)		
		+15 (oblique)		-38mm (measured)		

GHBMC 50th human body model

The GHBMC male 50th percentile model (version: FMB v.4.1.1, weight: 77.1 kg, and height: 175.3 cm, released Sep. 1st, 2013) was used in this study. This detailed FE human body model consists of 1.26 million nodes, 2.19 million elements and 981 parts. The model was developed in the seated driving posture, referred to as "UMTRI seating posture", with the arm angle at 41 degrees with respect to the horizontal plane to grab the steering wheel [14]; however, the arm of the PMHS in the tests was parallel to the upper torso. In order to match the experimental arm angle in the model, the arm of the GHBMC model was rotated by using the displacement control before conducting the impact simulation. In order to confirm that changing the arm angle did not have an effect on the other body parts, the bony structure of the model before and after rotating the arm was compared (Figure 1). In the PMHS tests, the forearms were crossed on the abdomen, however, this was not considered in this study due to the difficulty of matching that posture in the model. The angle of the thigh was fixed at 11 degrees for all test conditions in this study, since it was not the region of interest; additionally, the lower extremities moved far later than the shoulder due to inertia.

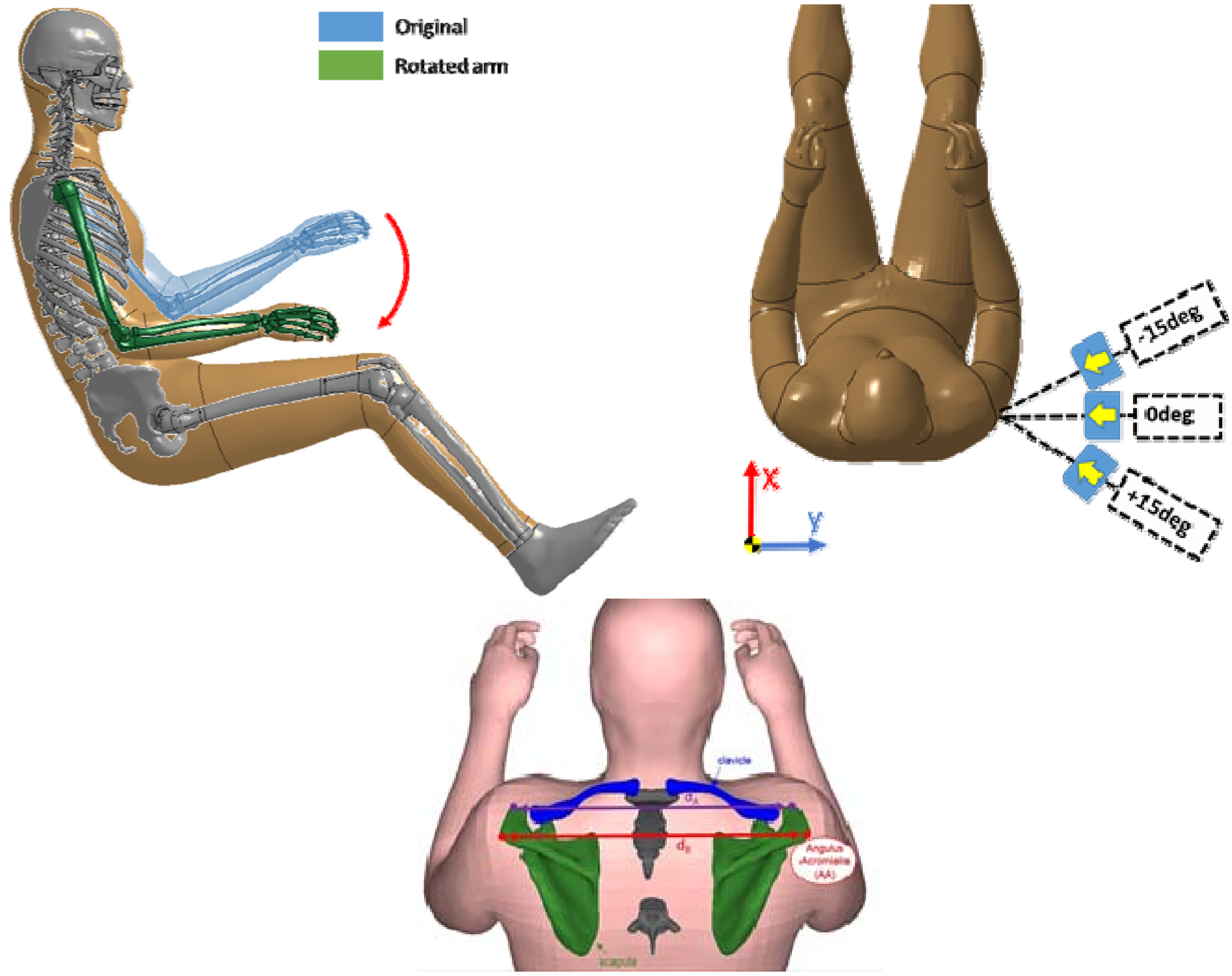

Fig. 1. GHBMC arm rotation to match the initial arm angle to that of PMHS (upper-left); impact direction of the model as well as the coordinate system used in this study (upper-right). The measurements of shoulder deflections for each test conditions (lower) : d_A on XY plane (Bolte's), d_A in three dimensional space (Compigne's), d_B in three dimensional space (Subit's)

Test Fixture in FE model

The seat was considered as a rigid body in the simulation. The friction coefficient between the rigid seat and the GHBMC model was defined as 0.249 to replicate the friction coefficient between the PMHS and the Teflon covered seat, as in the previous lateral impact simulation study [19]. A pre-simulation was conducted to consider the gravity on the body: the GHBMC model was located on the seat and gravity was applied for 1000ms, with the posture of the body constrained. Since the contact force between the GHBMC model and the seat converged, it was considered as the equilibrium state between the GHBMC model and the seat (Figure A2). The deformed geometry of the body during the pre-simulation was imported into the impact simulation, and the initial stress in the pelvis flesh of the model was included using the * INITIAL_FOAM_REFERENCE_ GEOMETRY option in LS-DYNA [16]. The deformed geometry of the model is shown in Figure A2.

In Subit's tests, the head was restrained by a wire connected to a spring (1.3 N/mm); the head was restrained in the FE model using a spring element of the same stiffness. During Bolte's test, the impacting ram was covered with a 5 cm thick piece of Arcel 310, 26.4 kg/m^3 density foam padding. In order to take into account the foam material characteristics, the foam was modeled using the *LOW_DENSITY_FOAM material model in LS-DYNA [16] using the pressure versus compression characteristic curve of this padding presented in Bolte et al. [11].

Impactor location, velocity, and stroke

In all three PMHS studies, the target location of impact was at the level of the glenohumeral joint. However, during the tests conducted by Subit et al., the actual impactor location was lower than the level of the glenohumeral joint in the vertical direction based on the analysis using the VICON data and CT scan image of the test (Table I). Thus, the location of the impactor in the FE model for Subit's test condition followed the impactor location indicated in Table I. For the simulations of Bolte's and Compigne's test conditions, the location of impactor was assumed at the level of the glenohumeral joint. The effect of impactor location relative to the level of glenohumeral joint in the vertical direction (-20mm, -40mm, -60mm) on the response of the shoulder of the GHBMC model was analyzed using Subit's test condition (Figure A1).

For the simulation of Bolte's and Compigne's test conditions, the mass of the impactor was matched by changing the material density of the part, and the initial speed was prescribed for the impactor. In other words, the kinetic energy was reduced after impacting the subject. In Subit's test, they used a relatively high mass (71.4kg) impactor to produce constant impacting speed and the stopper to limit the impact stroke as 80mm. For the simulation of Subit's test condition, the measured displacement of impactor from VICON data was directly used to prescribe the motion of the impactor.

Data Processing

All the data in this paper were presented according to the reference coordinate system in SAE J1733 (Figure 1) [15]. The shoulder impact force of the GHBMC model was taken from the normal component of contact force between the shoulder and the impactor along the impact direction. For comparison of the simulation results with the experimental data, the model's shoulder deflection was analyzed using the way of each study: the relative distance between the bilateral acromions on XY plane as in Bolte's study; the relative distance between the bilateral acromions in three-dimensional space as in Compigne's study; and the relative distance between the bilateral acromion angles (angulus acomialis, AA) in three-dimensional space as in Subit's study (Figure 1). The shoulder kinematics of the model was analyzed with respect to both the seat coordinate system and the T8 coordinate system used in Subit's study [13].

Bolte and Compigne scaled the PMHS responses from their tests to the standard 50th percentile male using the method developed by Mertz [21]. Subit et al., on the other hand, did not scale their data; so in the current study, the PMHS responses from Subit's tests were scaled in the same way as the other tests (eq. 1-3).

$$t_{scaled} = t_{subject} \times \sqrt{(75/m_{subject})/(468/l_{subject})} \quad (1)$$

$$D_{scaled} = D_{subject} \times \sqrt{(75/m_{subject})/(468/l_{subject})} \quad (2)$$

$$F_{scaled} = F_{subject} \times \sqrt{(75/m_{subject}) \times (468/l_{subject})} \quad (3)$$

where t = time, D = shoulder deflection, and F = impact force (subscript scaled = scaled response and subscript subject = subject response from the test), $m_{subject}$ = mass of the subject (kg), $l_{subject}$= shoulder width of the subject (mm). The force response from the model was filtered using the same filter class used in the PMHS tests (CFC180).

Since Bolte et al. [11] and Compigne et al. [12] showed that PMHS response is similar in left and right side impact tests, all the data in this paper were presented as the right side impact condition for the convenience of analysis. In other words, the impactor struck the right side of the GHBMC model in the FE model, and all experimental data from left side impacts were reflected about the sagittal plane and considered as right side impacts.

III. RESULTS

The comparison of the force-deflection response between the GHBMC model and the PMHS for the Bolte's test condition is presented in Figure 2. The GHBMC model predicted less shoulder deflection than the PMHS, but similar impact force response. The impact force and shoulder deflection time histories of the GHBMC model and PMHS for the Compigne's test condition are compared in Figure 3. In general, the impact force time history of the GHBMC model correlated well with the PMHS response, while the deflection of the model was less than the PMHS shoulder deflection from all tests in three impact directions.

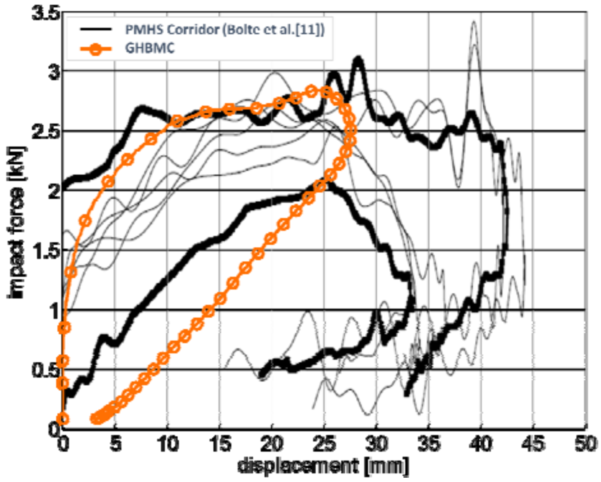

Fig. 2. Comparison of force-deflection response between the GHBMC model and the PMHS [11] at 4.4m/s impact along lateral direction (0°)

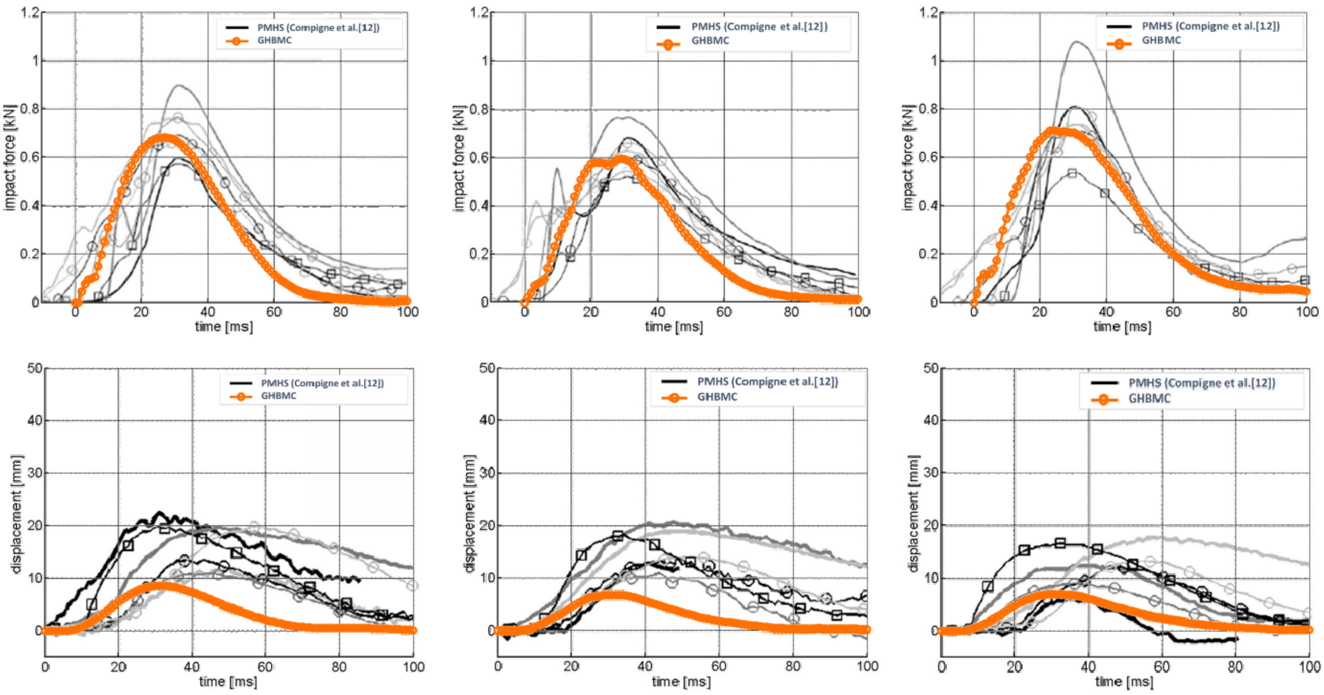

Fig. 3. Comparison of force time history between the GHBMC model and the PMHS [12] at 1.5m/s : 0°(left), -15°(middle), +15°(right) / Impact force (upper), deflection (lower)

The shoulder impact force time history, shoulder deflection time history and kinematics of the GHBMC model are compared to that of PMHS from Subit's tests in Figures 4-7. The GHBMC model predicted a similar shoulder impact force to that of PMHS from the different impact direction tests. On the other hand, the shoulder deflection of the model was different from the test response: the model only predicted a similar peak response for the 0° impact direction, and the model showed different response at the unloading phase. In terms of kinematics, the model predicted a similar motion to that of the experiments—shoulder compression along the impact direction, then upward excursion of the scapula—but the magnitude of the motion was less than that of the experiments.

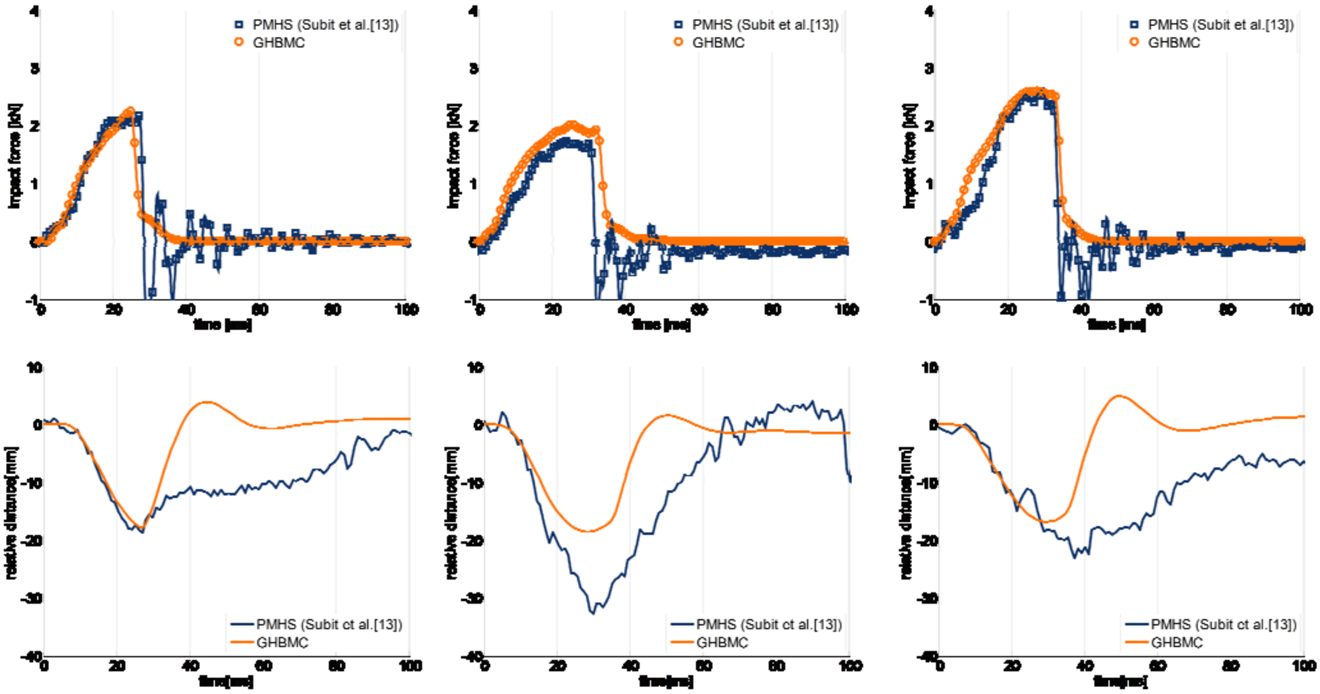

Fig. 4. Comparison of impact force (upper) and deflection (lower) time history between the GHBMC model and the PMHS [13] at 3.0m/s : 0° (left), -15° (middle), +15° (right)

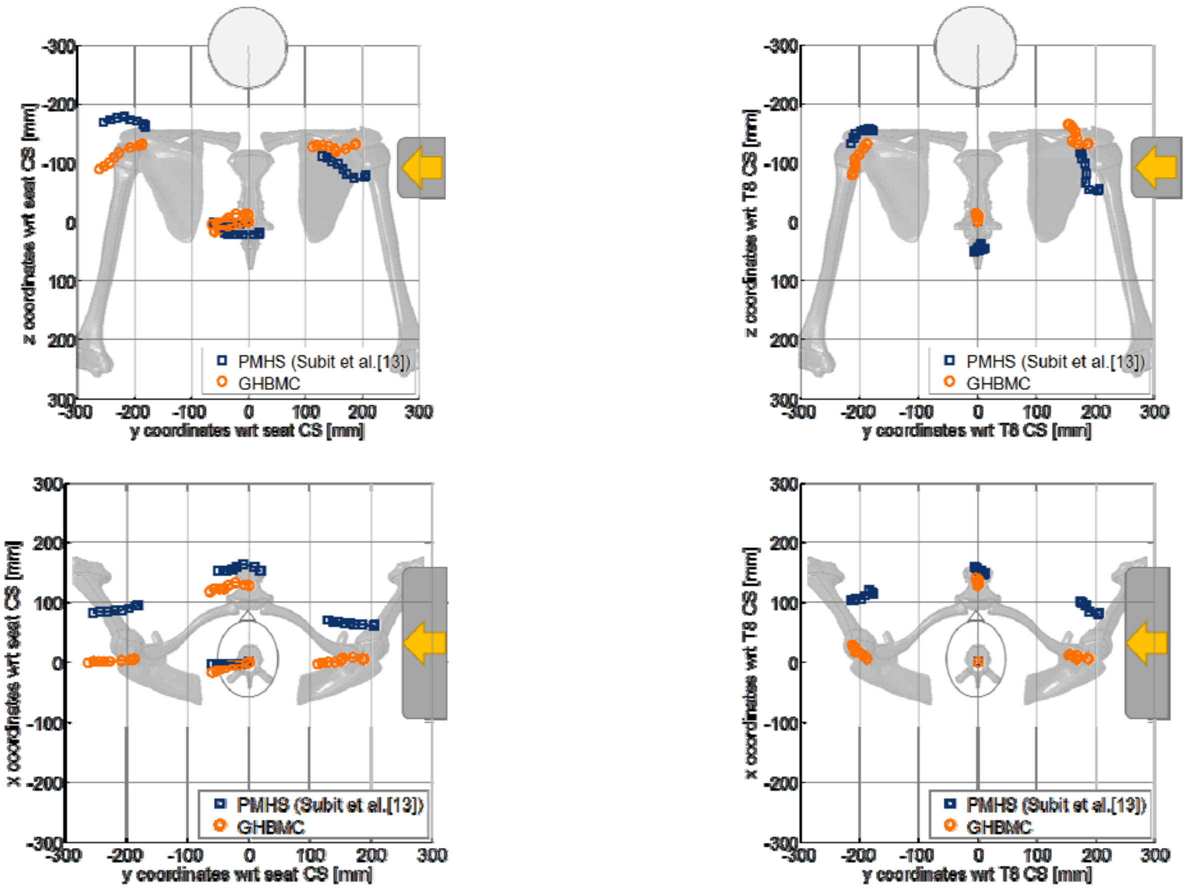

Fig. 5. Comparison of kinematics between the GHBMC and the PMHS [13] for 0° impact direction: Posterior view (upper), Superior view (lower) / with respect to seat (left), with respect to T8 (right)

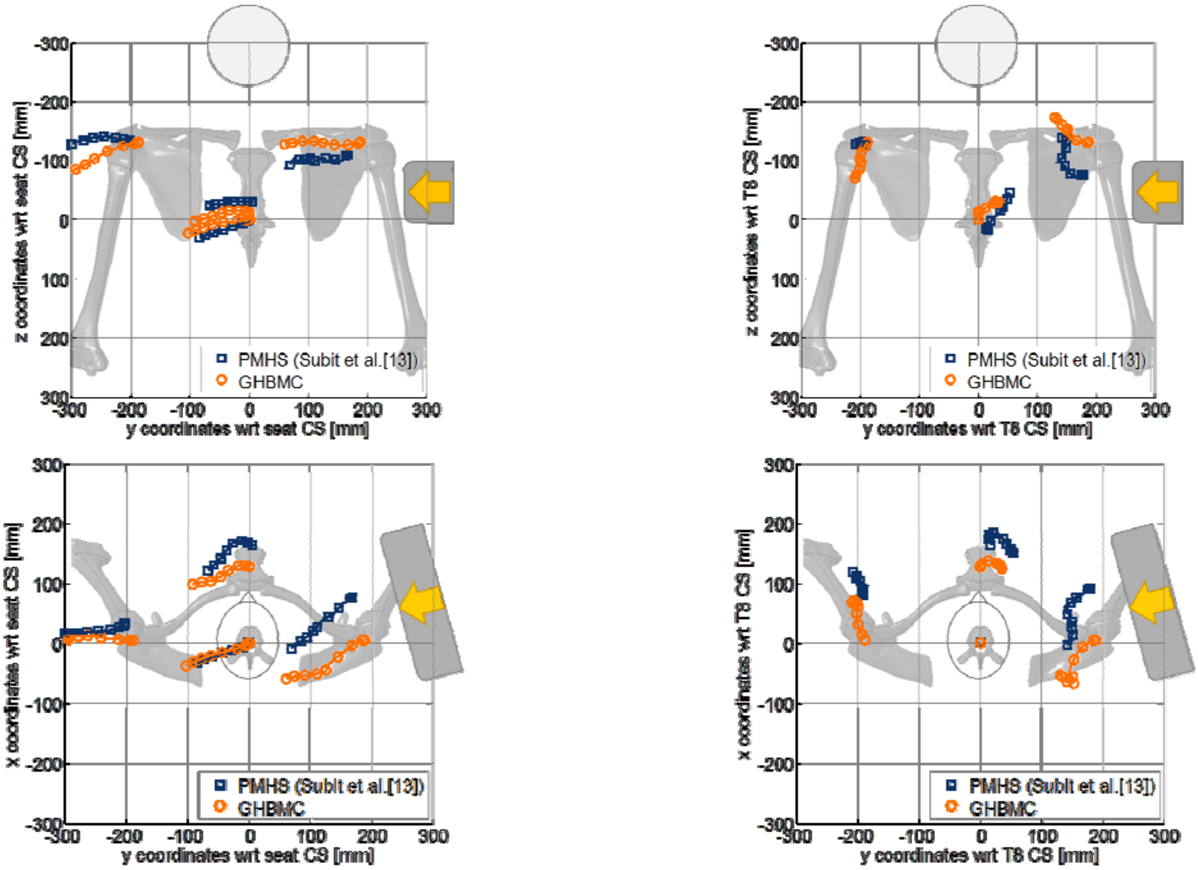

Fig. 6. Comparison of kinematics between the GHBMC and the PMHS [13] for -15° impact direction: Posterior view (upper), Superior view (lower) / with respect to seat (left), with respect to T8 (right)

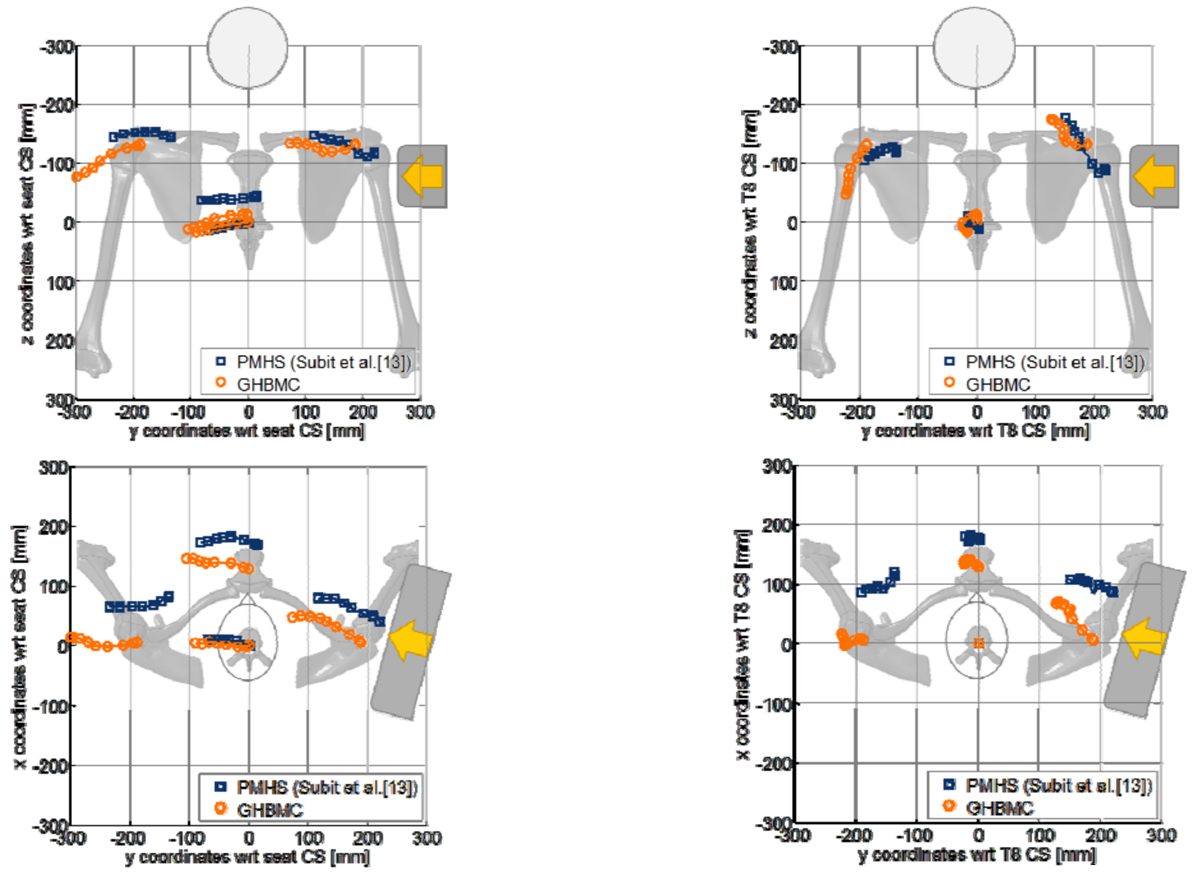

Fig. 7. Comparison of kinematics between the GHBMC and the PMHS [13] for +15° impact direction: Posterior view (upper), Superior view (lower)/with respect to seat (left), with respect to T8 (right)

IV. DISCUSSION

Impact force response of the GHBMC model

In general, the shoulder of the GHBMC model predicted a similar response to that of the PMHS in terms of impact force for the various loading conditions: thee impact speeds (1.5m/s, 3m/s, and 4.4m/s) and three impact directions (0^{o}, -15^{o}, and +15^{o}). In Compigne's test, the +15^{o} (posterolateral) impact resulted in a higher shoulder impact force than the other directions and the author claimed that the alignment of the impact direction with the clavicle could be one of the possible reasons [12]. The same trend was shown in Subit's test. The GHBMC model predicted the same shoulder impact force trend shown in the PMHS tests.

Shoulder deflection response of the GHBMC model – unloading response

The GHMBC model tended to predict lower peak shoulder deflection than those of PMHS (Figure 2, Figure 3, and Figure 4) and it failed to predict unloading phase of the shoulder deflection time histories of the PMHS. The shoulder deflection of the model was recovered quicker than those of the PMHS. This implies that the model dissipated less amount of energy stored during a loading phase. It is believed that the deflection of the shoulder is related to the ligaments and muscles that connect the bones in the shoulder region. Currently, there was no consideration of the viscoelasticity in the constitutive models used for the ligaments and muscles connected to the shoulder of the model. The viscoelasticity of those ligaments and muscles need to be taken into account in the GHBMC model to predict the unloading phase of the shoulder deflection time histories. Without proper modeling of the viscoelasticity of the shoulder region, it will be difficult to obtain biofidelic response under various impact speeds.

Shoulder deflection response of the GHBMC model – directional property

Compared to the results from Subit's tests with three impact directions, the model captured only the peak shoulder deflection in the 0^{o} impact direction. There are two possible reasons which can explain this. First, the discrepancy between the responses of the model and the PMHS may stem from subject variability such as anthropometry, initial posture, age and BMI, which were not considered in the simulations. In Subit's tests, the results from two subjects were included in this paper; the subject 427 is for the 0^{o} impact condition (subject 427) and the subject 420 for the -15^{o} and +15^{o} conditions (Table A1). Second, the predicted response for the oblique impact did not match the experimental data. In the PMHS tests, the shoulder deflection was greatest for the -15^{o} impact: the model does not capture this directional property of shoulder compliance. However, since the number of Subit's test included in this study is one for each impact direction, more shoulder kinematics data are needed from oblique impact tests to confirm the direction-dependent response of the shoulder.

Shoulder kinematics of the GHBMC model

The GHBMC model predicted a similar response to the PMHS response from Subit's test in terms of the general kinematics of the shoulder complex, which showed the compression along the impact direction followed by the upward displacement of the scapula. However, the model showed less scapular displacement than that of the PMHS for the oblique impact direction tests (-15^{o}, +15^{o}). Especially, comparing the right scapula kinematics with respect to T8 vertebra between the PMHS and the model in -15^{o} impact direction, the PMHS showed more scapular displacement toward anterior direction while the scapula motion was limited in the GHBMC model. This less scapular displacement with respect to the T8 resulted in the shoulder deflection difference between the test and the model.

Shoulder Injury Prediction using the GHBMC model

Since the injury risk functions of the shoulder use the shoulder deflection as the injury predictor [10-12], this study showed the limit of the GHBMC model in predicting shoulder injury. In other words, the model predicted less shoulder deflection than that of the PMHS for different test conditions, except for the 0^{o} impact in Subit's test. Furthermore, the model did not capture the directional property of the shoulder shown in Subit's and Compigne's tests: larger shoulder compliance in -15^{o} impact direction than 0^{o} and +15^{o} impact direction. This lack of prediction of shoulder deflection is likely related to the relative motion of the scapular to the other parts of the bodies.

Limitation of the study

The impact locations for Bolte's and Compigne's test conditions were assumed to be the level of the glenohumeral joint. However, due to the difficulty in controlling PMHS posture in the experiments, the impact location was likely to be different from the level of the glenohumeral joints as shown in Subit's tests. This assumption may prevent comparison of results between the model and the PMHS. Based on the analysis of the effect of impact location, it has an effect on the shoulder deflection response of the model up to 30% of its peak value (Figure A1). In order to perform a more precise evaluation, more detailed boundary conditions should be taken into account in the analysis as was done for the Subit's test condition. Also, the initial posture variance between the tests needs to be considered in the simulation. The kinematics analysis shown in Figure 5-7 showed the difference in initial posture of the subject between the PMHS and the GHBMC model: in the experiment, the scapula was anteriorly located with respect to T8 due to higher lordosis, and the entire torso was rotated along the X-axis; on the other hand, the GHBMC model was in a relatively upright seated posture. Even though the model could predict the response of the PMHS shoulder despite the different initial posture from the PMHS for the 0^o impact test in Subit's study, it would be better to match the initial posture between the model and the PMHS in order to minimize the difference between the model and the PMHS.

V. CONCLUSIONS

The biofidelity of the GHBMC shoulder was evaluated for various loading conditions: impact directions (0^o, -15^o, $+15^o$) and velocities (1.5, 3.0, 4.4 m/s). The impact force responses of the GHBMC shoulder showed a good correlation with the PMHS responses while the model predicted showed differences in the peak shoulder deflection and kinematics from those of PMHS. The GHBMC model predicted lower peak shoulder deformation and less amounts of energy dissipation than those of the PMHS. Since the shoulder deflection is used to predict injury risk of the shoulder complex, the GHBMC model needs to be improved to predict the peak shoulder deflection amounts. Since it is believed that the shoulder deflection is closely related to the properties of the ligaments and the muscles in the shoulder complex, further validation of those soft tissue in the shoulder complex is required.

VI. REFERENCES

[1] Beillas P, and Berthet F. Performance of a 50th percentile abdominal model for impact: effect of size and mass. *European Society of Biomechanics Conference*, 2012.

[2] DeWit JA, and Cronin DS. Cervical spine segment finite element model for traumatic injury prediction. *J. Mech. Behav. Biomed. Mater*, 2012, 10:138–150.

[3] Fice JB, Cronin DS, and Panzer MB. Cervical spine model to predict capsular ligament response in rear impact. *Ann. Biomed. Eng.*, 2011, 39:2152–2162.

[4] Li Z, Kindig MW, et al. Rib fractures under anterior-posterior dynamic loads: experimental and finite element study. *J. Biomech.*, 2010, 43:228–234.

[5] Li Z, Kindig MW, Subit D and Kent RW. Influence of mesh density, cortical thickness and material properties on human rib fracture prediction. *Med. Eng. Phys.*, 2010, 32, 998–1008.

[6] Mao H, Zhang L, et al. Recent advances in developing finite element head model, *International Crashworthiness Conference*, Milan Italy, 2012.

[7] Shin J, Yue N., and Untaroiu CD. A finite element model of the foot and ankle for automotive impact applications. *Ann. Biomed. Eng.*, 2012, 40(12):2519-31.

[8] Park G, Kim T, et al. Evaluation of biofidelity of side impact computational surrogates (ES-2re, WorldSID, GHBMC). 2014, SAE Technical Paper 2014-01-0541.

[9] Irwin AL, Walilko TJ, Cavanaugh JM, Zhu Y, King AI. Displacement responses of the shoulder and thorax in lateral sled impacts. *Proc. 37th Stapp Car Crash Conference*, 1993, Warrendale, PA.

[10] Koh SW, Cavanaugh JM, Zhu J. Injury and response of the shoulder in lateral sled tests. *Stapp Car Crash Journal,* 2001, 45: 101-141.

[11] Bolte JH 4th, Hines MH, Herriott RG, McFadden JD, Donnelly BR. Shoulder impact response and injury due to lateral and oblique loading. *Stapp Car Crash Journal*, 2003, 47:35-53.

[12] Compigne S, Caire Y, Quesnel T, Verries JP. Non-injurious and injurious impact response of the human shoulder three-dimensional analysis of kinematics and determination of injury threshold. *Stapp Car Crash Journal*, 2004, 48:89-123.

[13] Subit D, Duprey S, Lau S, Guillemot H, Lessley D, Kent R. Response of the human torso to lateral and oblique constant-velocity impacts. *Annals of Advances in Automotive Medicine,* 2010, 54:27-40.

[14] Schneider LW, Robbins DH, Pflug MA, Snyder RG. Anthropometry of Motor Vehicle Occupants, Vol. 3. Specifications and Drawings. Report HS-806 717, UMTRI-83-53-2, UMTRI, 1983.
[15] Society of Automotive Engineers. SAE Surface Vehicle Information Report: Sign Convention for Vehicle Crash Testing – SAE J1733. 1994.
[16] LS-DYNA Keyword User's Manual, Version 971, *Livermore Software Technology Corporation.*
[17] Eppinger R, Marcus J, Morgan R. Development of dummy and injury index for NHTSA's thoracic side impact protection research program, 1984, SAE Technical Paper 840885.
[18] Bolte JH 4th, Hines MH, McFadden JD, Saul RA. Shoulder response characteristics and injury due to lateral glenohumeral joint impacts, 2000, SAE Technical Paper 2000-01-SC18.
[19] Park G, Kim T, Crandall JR, Carlos AD, Javier LN. Comparison of Kinematics of GHBMC to PMHS on the Side Impact Condition, *Proceedings of IRCOBI Conference*, 2013, Gothenburg, Sweden.
[20] LS-DYNA Keyword User's Manual, Version 971, Livermore Software Technology Corporation
[21] Mertz, HJ. A procedure for Normalizing Impact Response Data. SAE Technical Paper 840884, 1984, Warrendale, PA.

VII. APPENDIX

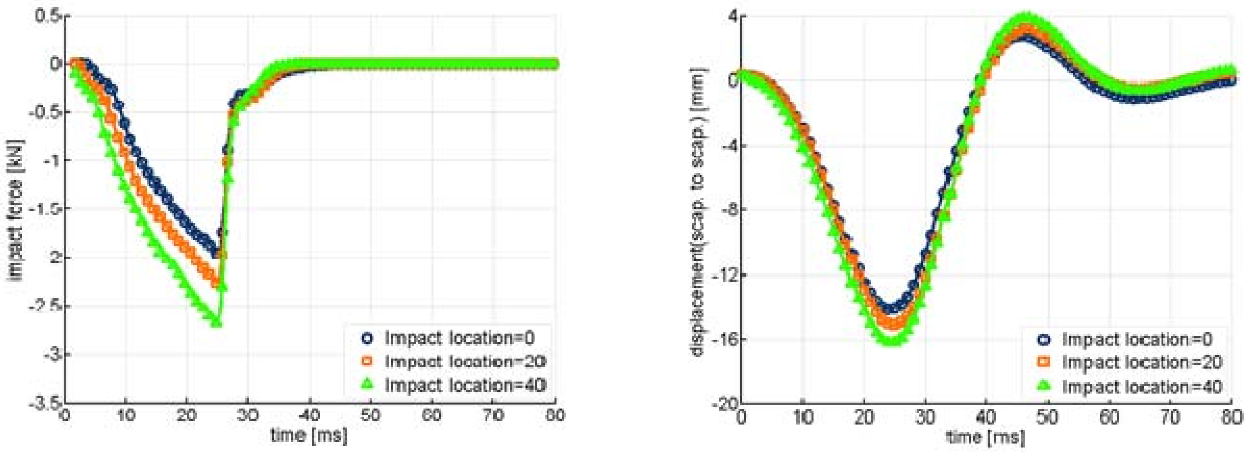

Fig. A1. Effect of impactor location with respect to the level of glenohumeral joint on the shoulder response

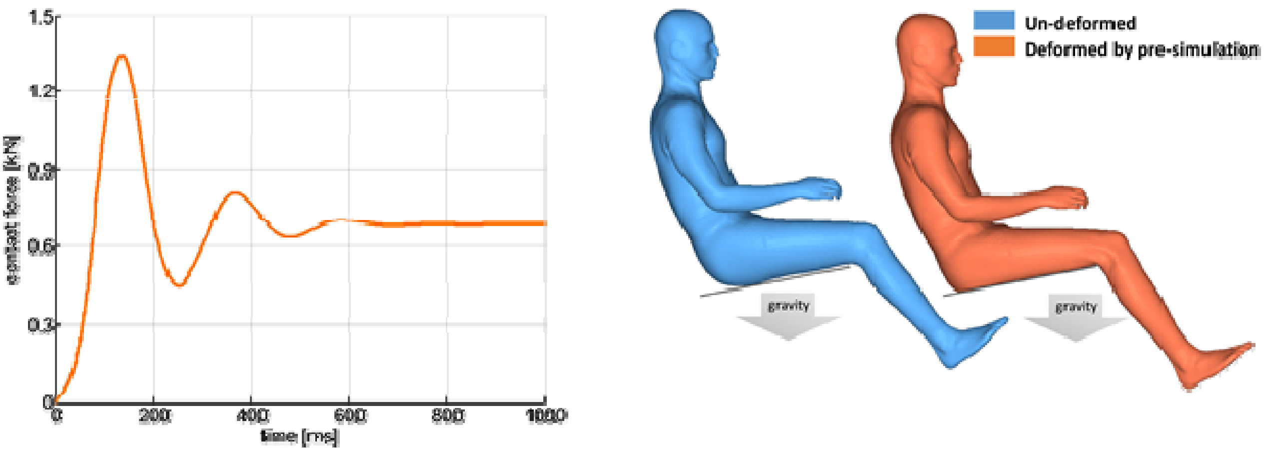

Fig. A2. Contact force during the pre-simulation (left) and the deformed geometry after the pre-simulation (right)

Table A1: Summary of the subject characteristics from Subit's test [13]

	Subject ID	Age (year)	Weight (kg)	Stature (cm)	BMI (kg/m2)
0^{o}	427	79	79	181	24.1
$-15^{o}/+15^{o}$	420	59	93	180	28.7