Head Kinematics and Shoulder Biomechanics in Shoulder Impacts Similar to Pedestrian Crashes—A THUMS Study

RUTH PAAS, JOHAN DAVIDSSON, and KARIN BROLIN
Applied Mechanics, Chalmers University of Technology, Gothenburg, Sweden

Received 26 March 2014, Accepted 19 September 2014

Objective: Head injuries account for the largest percentage of fatalities among pedestrians in car crashes. To prevent or mitigate such injuries, safety systems that reduce head linear and rotational acceleration should be introduced. Human body models (HBMs) are valuable safety system evaluation tools for assessing both head injury risk and head kinematics prior to head contact. This article aims to evaluate the suitability of the Total Human Model for Safety (THUMS) version 4.0 for studying shoulder impacts, similar to pedestrian crashes, investigating head, spine, and shoulder kinematics as well as shoulder biomechanics.

Methods: Shoulder impact experiments including volunteers and postmortem human subjects (PMHSs) were simulated with THUMS. Head linear and angular and vertebral linear displacements of THUMS were compared with volunteers and shoulder deflections with both volunteers and PMHSs. A parameter variation study was conducted to assess head response to shoulder impacts, by varying shoulder posture and impact directions mimicking shoulder-to-vehicle contacts. Functional biomechanics literature was compared with THUMS responses in view of pedestrian-like shoulder impacts.

Results: THUMS head linear displacement compared better with tensed than with relaxed volunteers. Head lateral rotation was comparable with volunteer responses up to 120 ms; head twist was greater in THUMS than in the volunteers. The THUMS spine appeared to be stiffer than in the volunteers. Shoulder deflections were smaller than in the relaxed volunteers but matched the PMHSs. Raised shoulder postures decreased the THUMS shoulder deflections and increased head lateral displacements. When the impactor surface orientation or the impact velocity angle was changed from lateral to superolateral, THUMS head lateral displacement decreased. THUMS scapula and clavicle kinematics compared well with previous experimental studies. The shoulder impact conditions influenced the scapula motion over the thorax, which had considerable effect on upper torso and head kinematics. The clavicle primarily acted as a guide for the scapula. In the PMHS experiments, it took 20 ms from first impactor-to-shoulder contact to head response, indicating that shoulder impacts in pedestrian crashes may influence head kinematics during head impact.

Conclusions: THUMS is generally suitable for studying head linear kinematics and head lateral rotation in shoulder impacts similar to pedestrian crashes and for studying shoulder girdle biomechanics. Head twist and spine stiffness were more pronounced than in the volunteers. The results have identified the need for additional volunteer shoulder impact testing, mimicking pedestrian crashes, as well as the need to address shoulder impacts in full-scale pedestrian experiments.

Keywords: pedestrian, head, shoulder, spine, THUMS, biomechanics

Introduction

Head injuries are the leading cause of death in pedestrian-to-vehicle crashes (Lau et al. 1998). Elbow and shoulder impacts in pedestrian crashes have considerable effect on head kinematics (Paas et al. 2012). For the development of pedestrian safety systems, it is necessary to further study the influence of elbow and shoulder impacts on head kinematics. Common tools to assess pedestrian safety systems are human body models (HBMs), such as the commercially available, full-body, finite element HBM THUMS (Toyota Motor Corporation 2011). In contrast to currently available multibody models, detailed finite element HBMs enable studying shoulder–girdle biomechanics, including tissue deformation.

The pedestrian THUMS version 4.0 has been evaluated to some extent at component level and against full-scale pedestrian tests by Shigeta et al. (2009) and Watanabe et al. (2011, 2012), as listed in Appendix C1 (see online supplement). They evaluated the humerus with static 3-point bending and dynamic compression (Kemper et al. 2005). In their pedestrian evaluations, 2-dimensional head, T1, and L5 linear displacements were evaluated against 3 experiments. The head impact location was predicted within 100 mm. No vehicle model validation was found in the literature. For pedestrian head injury prediction, THUMS should provide biofidelic 6-dimensional head kinematics, including elbow and shoulder
impacts against a validated vehicle model. To the best of the authors’ knowledge, evaluation of the THUMS 4.0 shoulder, spine, and head responses to shoulder impacts has not been published to date.

Human responses to shoulder impacts have been examined in several experimental studies. Experiments in which only the shoulder was impacted were conducted by Bolte et al. (2000, 2003), Thollon (2001), Marth (2002), Compigne et al. (2004), and Subit et al. (2010) using postmortem human subjects (PMHSs) and by Ono et al. (2005) using volunteers. In these studies, the shoulder was impacted laterally and, in some cases, anterolaterally and posterolaterally. The authors’ investigations have found that only the studies carried out by Bolte et al. (2003) and Ono et al. (2005) ensured that the shoulders were not in an inferior position prior to impact. Side impact responses have been investigated in a number of studies where multiple body parts were impacted at the same time, in addition to the shoulder, such as Cavanaugh et al. (1990), Irwin et al. (1993), and Koh et al. (2001).

Shoulder impact conditions during pedestrian experiments are different from the aforementioned shoulder experiments. Shoulder impacts with the bonnet or windshield occurred in 10 of 13 pedestrian experiments in Paas et al. (2012), Schroeder et al. (2008), Kerrigan, Kam, et al. (2005), Kerrigan, Murphy, et al. (2005), and Kerrigan et al. (2007) based on the available video footage. In 7 of the 13 experiments an elbow impact occurred, causing shoulder elevation or elevation and anterior displacement immediately before shoulder impact. The orientation of the impact surface and the direction of the resultant impact velocity varied depending on vehicle velocity and geometry. Experimental or numerical studies investigating shoulder impacts similar to those in pedestrian crashes could not be found in the literature.

The purpose of this study is to evaluate head, spine, and shoulder kinematics of THUMS version 4.0 in pedestrian-like shoulder impacts. THUMS is compared with volunteer and PMHS experiments in lateral, anterolateral, and posterolateral shoulder impacts. A THUMS parameter variation study is conducted to examine head kinematics in different initial shoulder postures and impact directions that are likely to occur in pedestrian crashes. Functional biomechanics literature is discussed in comparison with THUMS responses in shoulder impacts similar to those in pedestrian crashes.

Methods

Shoulder impact experiments with volunteers (Ono et al. 2005) and with PMHSs (Bolte et al. 2003) were reanalyzed and simulated with THUMS 4.0 (Toyota Motor Corporation 2011). In addition, a parameter variation study was conducted, assessing the shoulder response in different initial shoulder postures (neutral, elevated, or anteriorly elevated), impact directions (90 or 45°), and impactor orientations (0 or 30°), in order to investigate shoulder and head responses in impact conditions similar to pedestrian full-scale experiments. Additional information on the parameter variation study can be found in Appendix A (see online supplement).

Fig. 1. Simulation setups in the evaluation study and coordinate system used throughout the study. Top: volunteer evaluation study with lateral (left), 15° anterior (middle), and 15° posterior impact (right). Bottom: PMHS evaluation study in frontal, side, and top views.

Experimental Data

In Ono et al. (2005), 8 volunteers (Appendix B, see online supplement) were impacted against the glenohumeral joint, fully lateral and 15° oblique in anterior and posterior directions, by a flat plate that was rigidly connected to an air-driven piston via a force transducer. The air pressure was adjusted to produce peak contact forces of 600 N, creating an impact–velocity curve resembling half a sinusoidal wave lasting 73 ms with an average peak velocity of 1.5 m/s after 45 ms. Volunteers were seated with their backs straight on a stool. Before each test, they were instructed either to relax or to tense their muscles. Kinematics were obtained with 3-dimensional motion tracking (Vicon).

In Bolte et al. (2003), 3 PMHSs (Appendix B) were impacted laterally against the glenohumeral joint by a 23-kg guided and padded impactor (5-cm-thick Arcel 310, 26.4 kg/m³ density) at an initial speed of 4.4 m/s. The PMHSs were positioned on a Teflon-coated impactor seat with a backrest and arm support (Figure 1). PMHS kinematics were obtained with accelerometer data.

The original volunteer Vicon data and PMHS accelerometer data were reanalyzed to provide displacement corridors (average ± standard deviation of the displacements). Volunteer spinal bending was calculated based on the available vertebral displacements from the Vicon data from C3 to L3. To provide bending deflections of the whole spine, C1 and L5 displacements were extrapolated from cubic spline interpolations through the vertebral, head center of gravity (CG), and pelvis displacements.

Evaluation Study

Simulations were carried out with THUMS version 4.0 in LS-DYNA version 971, release 6.1.0 (LSTC Inc., Livermore, CA) and postprocessed with LS-PREPOST (LSTC Inc.).
The pedestrian THUMS was used to simulate the volunteer experiments. Its legs were removed and the most inferior pelvic flesh was fixed in space; the movement of bones was not restricted directly. The impactor (150 mm × 100 mm) was modeled with hex elements and a linear elastic material model (Young’s modulus = 70 GPa, Poisson ratio = 0.3). The velocity was prescribed as measured in the experiments to a rigid shell on the rear of the impactor model.

The occupant THUMS was used for simulation of the PMHS experiments, because its posture was close to the PMHSs’. Minor differences in contact and material definitions (Appendix C2, see online supplement) were adapted to the pedestrian version. A rigid seat (450 mm × 500 mm, 16° angle; backrest 500 mm × 911 mm, 105° angle) made of quadratic shell elements, including a footrest, and near-rigid arm supports of hex elements (Figure 1) were fixed in space. The impactor was modeled as a rigid shell (200 mm × 150 mm) with a mass of 23 kg, covered with 50-mm low-density foam (Young’s modulus = 5.13 MPa) hex elements and low-density elastic shell elements. The impactor was given an initial velocity of 4.4 m/s.

For positioning before all simulations, force was applied to the THUMS wrists and elbows to make the upper and lower arm angles match the pre-impact photos. The spine curvature of the THUMS pedestrian matched pre-impact Vicon data of the volunteer spines. Before the PMHS experiment simulations, gravity was applied to the occupant model until it made contact with the seat and footrest; the head, neck, and upper body contours then appeared to correspond to the PMHS pre-impact photos. Friction coefficients of 0.3 were used for contacts between THUMS and the impactor, chair, and armrests. Data outputs were contact forces between the impactor and THUMS, nodal data from the acromia, the head CG, the sternum upper end, and the vertebral body centers of C1, C3, T1, T2, T4, T5, T8, T9, T12, L1, L3, and L5.

Throughout the study, in the experiments and simulations, the coordinate system (Figure 1) was defined according to the Society of Automotive Engineers (SAE 1994). Head linear displacements were defined as head CG linear displacements. Head angular displacements were calculated as successive Euler angles in the order x-y′-z″; that is, the head first rotated around the anteroposterior x-axis, then around the new y-axis, and finally around the new z-axis. Lateral rotation was defined as angular x-displacement and twist as angular z-displacement. Spine lateral bending was defined according to the beam bending theory, as the bending deflection of the vertebrae compared with a straight line through the vertebral body centers of C1 and L5. Shoulder deflections were defined as changes in absolute distances between the impacted acromion and T1 (acromion-to-T1 deflection) or sternum (acromion-to-sternum deflection). All experimental and simulation data were nonscaled and filtered with a CFC 60 (Appendix F, see online supplement) low-pass Butterworth filter (SAE 1995).

**Results**

In the first part of this section, THUMS results are evaluated against the volunteer and PMHS data. Subsequently, the main results of the parameter variation study are presented (detailed results in Appendix A).

**Evaluation Study**

Head linear displacements of THUMS compared better to those of the tensed than those of the relaxed volunteers (Figure 2). Compared to the relaxed volunteers, the magnitude of THUMS head y-displacement was outside the corridors for all impact directions and approximately twice as large at 150 ms. Compared with the tensed volunteers, THUMS head y-displacement was within the corridor in the lateral impact, slightly greater in the anterior impact, and slightly shorter in the posterior impact. Forward and downward head linear displacement results are superimposed on relaxed (dashed lines) and tensed (dotted lines) volunteer corridors. Lateral (left), 15° anterior (middle), and 15° posterior impacts (right).

![Fig. 2. Head y-displacements of THUMS (continuous lines) superimposed on relaxed (dashed lines) and tensed (dotted lines) volunteer corridors. Lateral (left), 15° anterior (middle), and 15° posterior impacts (right).](image-url)
displacements were generally low in all simulations and volunteer experiments (Appendix E2, see online supplement). Head twist was considerably greater in THUMS than in the volunteers, whereas head lateral rotation remained within or close to the upper boundary of the volunteer corridors until 120 ms. Head $\gamma$-rotations were consistently below 3$^\circ$ for all simulations and volunteers (Appendix E1, see online supplement).

In the volunteer simulations, THUMS exhibited spine lateral bending after 30 ms (Figure 3). The maximum lateral bending was observed at the T4/T5 level with a bending deflection of 21 mm at 80 ms. Afterwards, the location of maximum bending gradually moved down toward T8 and decreased to 10 mm at 150 ms. In the average relaxed volunteer, spine lateral bending occurred after 40 ms. Maximum lateral bending in the relaxed volunteer spine occurred at T1/T2 level with a bending deflection of approximately 38 mm at 120 ms. This indicates that the THUMS spine was stiffer and started to return to initial curvature earlier than that of the relaxed volunteers. Although the THUMS spine moved slightly less overall than the relaxed volunteers, the smaller magnitude and inferior location of the spinal bending resulted in greater head $\gamma$-displacement than in the relaxed volunteers. In the average tensed volunteer, maximum lateral spinal bending occurred at the T4/T5 level with a bending deflection of 24 mm at 95 ms. Compared to the tensed volunteers, the THUMS spine moved less and the curvature was slightly less pronounced. This produced a close match of the head $\gamma$-displacements between THUMS and the tensed volunteers. Head $\gamma$-displacements were influenced by the combination of spine curvature and spine overall lateral motion.

Peak acromion-to-sternum and acromion-to-T1 deflections in THUMS were consistently smaller than in the average relaxed volunteer (Figure 4), indicating stronger coupling between the shoulder and the spine in THUMS compared to the relaxed volunteers. Compared to the average tensed volunteer, peak acromion-to-sternum deflections in THUMS were of similar magnitude. Peak acromion-to-T1 deflections were greater in THUMS than in the average tensed volunteer in the lateral and posterior impact (indicating weaker coupling) and smaller in the anterior impact (indicating stronger coupling between the shoulder and spine). In all impact directions, THUMS peak acromion-to-T1 deflections occurred later than in the tensed volunteers. After the impactor stopped,
the deflections generally decreased faster in both the relaxed and tensed volunteers than in THUMS. During shoulder rebound, the volunteers exhibited acromion-to-sternum and acromion-to-T1 distances that were greater than the original distances—that is, positive deflections—which was not observed in THUMS. The standard deviations were relatively large; THUMS shoulder deflections generally remained within both the relaxed and the tensed volunteer corridors. In summary, THUMS shoulder deflections generally compared better with the tensed than with the relaxed volunteers in the first 70 ms of the impact. If the scapula-to-spine coupling were weakened with the tensed rather than the relaxed volunteers in the first 70 ms of the impact, it is expected that the spine curvature and head displacements would be reduced.

The acromion-to-T1 deflection of THUMS compared well with the PMHSs (Figure 5), both in peak magnitudes and in peak timing. The peak acromion-to-sternum deflection was smaller in THUMS than in the average PMHS, although THUMS remained inside the corridor, with relatively large variability in the experimental data. The THUMS shoulder did not rebound as fast as the PMHSs’; THUMS acromion-to-T1 deflections were outside the corridor after approximately 40 ms. Sternum and T1 displacements of THUMS closely agreed with the average PMHS response and remained within the PMHS corridors until approximately 90 ms after first contact (Appendix E4, see online supplement).

**Parameter Variation Study**

Head y-displacements increased when changing the shoulder posture from neutral to elevated and anteriorly elevated positions (Appendix A, Figure A-3, see online supplement). In contrast, head x-displacements decreased when the impact direction was changed from lateral to superolateral (Figure A-3). The shoulder posture generally influenced head y-displacements more than the impactor configuration. Shoulder deflections indicated that the elevated and anteriorly elevated shoulder postures increased the coupling of shoulder and spine compared to the neutral shoulder posture (Appendix A).

**Discussion**

This article evaluated head, spine, and shoulder kinematics of THUMS version 4.0 in view of pedestrian crashes. This section discusses the effect of muscle response in HBM evaluations, the use of experimental data, shoulder girdle biomechanics, and implications for pedestrian crashes.

**Effect of Muscle Response**

Human subjects present different responses due to muscle activity. Jarrett and Saul (1998) reported that approximately 60% of pedestrians were unaware of the imminent impact, but a considerable percentage showed avoidance reactions such as bracing or jumping. Therefore, the development of 2 versions of THUMS is suggested to study the influence of muscle tension in pedestrians. A pedestrian aware of an imminent impact might tense the muscles strongly, more than a volunteer instructed to tense his muscles would. An unaware, relaxed pedestrian activates muscles for posture maintenance, whereas PMHSs completely lack muscle tonus. The mechanical responses of all human subjects vary depending on individual factors such as age, anthropometry, or bone mineral density. Due to these differences, the actual curves were compared for enhanced understanding of human and THUMS responses. In the future, THUMS may be fitted with active muscles to simulate pedestrians who are bracing prior to impact; this function has been implemented in a version of the occupant THUMS (Osth et al. 2012). In the present study, THUMS compared better with the tensed than with the relaxed volunteers, indicating that some muscle stiffness is included in the overall stiffness of the model. This should be taken into account before incorporating active muscles in THUMS.

**Inclusion of Experimental Data**

Only Ono et al. (2005) and Bolte et al. (2003) matched the focus of this study, which was on shoulder evaluation rather than on combined shoulder and upper torso evaluation in noninjurious shoulder impacts. In these studies, the impactor was directed against the glenohumeral joint and the thorax was not hit by the impactor; major thorax deformation was not observed. The shoulders were in a neutral position due to muscle force in the volunteers and extra arm support for the PMHSs. THUMS has not yet been validated for shoulder injury, thereby excluding Bolte et al. (2000). Cavanaugh et al. (1990), Irwin et al. (1993), and Koh et al. (2001) were excluded due to multiple load paths. Other studies—for example, Compigne et al. (2004) and Subit et al. (2010)—were excluded because the upper arms of the PMHSs were not supported during the test. The subjects’ shoulders were then likely to be in an inferior position during impact, changing the load path direction toward the thorax rather than the shoulder girdle.

**Limitations of the Experimental Data**

Different limitations were observed in the data collected from volunteers and PMHSs. In Ono et al. (2005), the Vicon
targets were attached to the skin of the volunteers; the targets did not oscillate visibly on the skin although the bones may have moved slightly differently than the skin. The number of volunteers was fair \((n = 8)\), and the boundary conditions were well determined. In Bolte et al. (2003), instrumentation was attached to the bones of the PMHSs, but the exact locations and orientations could not be determined from the available data. The number of PMHSs was low \((n = 3)\). The angles of the upper and lower arms were reproduced with an estimated error of below 10°; the results were not sensitive to variation of the arm angles \((±10°\text{ arm elevation})\) and the spine curvature (occupant or pedestrian). The seat properties in the oblique load cases were unknown and influenced the results considerably; these load cases were excluded.

**Shoulder Impact Severity**

Because head displacements were not available in Bolte et al. (2003), lower-severity \((1.5\text{ m/s})\) volunteer impacts (Ono et al. 2005) were used for evaluation of head kinematics. From previous full-scale pedestrian studies at a vehicle speed of 40 km/h (Kerrigan, Kam, et al. 2005; Kerrigan, Murphy, et al. 2005; Kerrigan et al. 2007; Paas et al. 2012), the resultant shoulder impact velocity against a small sedan was estimated to be around 10 m/s, against a mid-sized sedan around 7 m/s, and against an SUV around 7.5 m/s, thus considerably higher than in the volunteer evaluation. In pedestrian-to-vehicle simulations with varying vehicle models and impact speeds \((20–50\text{ km/h})\), Watanabe et al. (2012) reported unfiltered peak shoulder impact forces of 0.5–9.5 kN and unfiltered peak head impact forces of up to 30 kN. Kerrigan et al. (2012) measured pedestrian dummy head impact forces, averaged over HIC15 calculation time, of only 1–3.2 kN at 40 km/h. This discrepancy could be caused by vehicle model stiffness or impacts with stiff vehicle components in Watanabe et al. (2012).

The authors of the present study believe that shoulder impact forces tend to be considerably lower than head impact forces and that shoulder impact forces of 600 N may be common in 40 km/h pedestrian impacts, for 2 reasons: Firstly, video footage from previous full-scale pedestrian experiments indicates that the shoulder tends to roll off, resulting in a longer impact duration and lower peak force than in typical impactor scenarios (without rolling off). Secondly, shoulder impacts with vehicles are usually composed of 2 deformable surfaces, reducing peak contact force. The shoulder can absorb a greater amount of energy than the head without being injured, due to the large amount of surrounding soft tissue and its large range of motion during deformation. The prescribed sinusoidal velocity curve in the volunteer simulations replicated a deformable impact surface to some extent.

In real pedestrian crashes, shoulder impact conditions can vary considerably, due to a wide range of vehicle speeds, front shapes, and stiffness, producing different shoulder impact speeds and forces. Before using THUMS for real-life pedestrian safety evaluations, validating THUMS against volunteer responses is required. Ono et al. (2005) provided head and spine kinematics of relaxed and tensed volunteers, which could not be found in any other publication. Bolte et al. (2003) enabled studying shoulder response at higher severity \((4.4\text{ m/s, 23 kg})\). Therefore, these 2 studies provide valuable data for assessing shoulder impacts in pedestrian crashes.

**Functional Shoulder Girdle Biomechanics**

In humans, the shoulder girdle couples the upper extremities to the axial skeleton. The clavicle provides the only bony connection. The scapula is connected to the ribs and spinous processes via muscles (Marieb and Hoehn 2010), representing a physiological articulation between the scapula and the posterior thorax. These muscles control the scapular motion over the rib cage (Kibler 1998; Paine and Voigh 2013). Moseley et al. (1992) and Kamkar et al. (1993) studied the function of the largest muscle groups in scapular movements during shoulder rehabilitation exercises, concluding that these muscles play a vital role in shoulder movement and stabilization. In shoulder impacts, muscles influence the scapular coupling to the rib cage and spine through stabilization and damping, which was observed accordingly in THUMS. The largest muscle groups contributing to the scapular motion (Kibler 1998) were all modeled in THUMS as spring and solid elements.

Scapular response during THUMS positioning in elevated and anteriorly elevated shoulder postures appeared to be human-like. Upward displacement of the THUMS humerus (shrugging) caused scapular elevation, medial displacement, and upward rotation by 30.7°, just below the reported maximum for the human range of motion \((36.8° \pm 5.0°\text{; Hallac¸eli and G ¨unal 2002})\). Similar scapular motion has been observed during humerus abduction (Ludewig et al. 1996). In THUMS, the scapular upward rotation caused an elevation of the clavicle, similar to human shoulders (Kamkar et al. 1993). During positioning of the THUMS shoulder in the anteriorly elevated posture, the scapula was elevated, rotated upwards, and protracted. Scapular protraction consists of lateral sliding and forward tilting around the thorax (Weisner et al. 1999), as observed in THUMS.

**Shoulder Girdle Impact Response**

The shoulder-to-spine coupling in shoulder impacts can be described as the proportion of the inserted energy transmitted to the spine. Based on the functional biomechanics literature and the simulation results, this coupling appeared to be governed by the interaction of the scapula with the muscles connecting the scapula with the ribs and spine.

Figure 6 displays the scapular motion in the lateral, anterior, and posterior impacts. Although load transfer contributions from the clavicle and surrounding tissue were observed in the simulations, these appeared to be secondary. For clarity, they are omitted in the following reasoning but discussed further below. The (simplified) load path from shoulder impact to spine motion appeared to be as follows: The impactor pushed the soft tissue of the shoulder medially, causing the head of the humerus to translate medially toward the glenoid cavity. Bony contact and contributions from the surrounding soft tissue caused scapular medial sliding over the rib cage and
rotation depending on the impact direction. During sliding, the scapula was pressed into the subscapularis muscle tissue, which was pressed into the rib cage. The ribs transferred the load to the thoracic spine via the costal facets. The combination of scapular medial sliding and pressing the subscapularis into the rib cage appeared to be the primary load path to the spine, governing the shoulder-to-spine coupling.

The scapular medial sliding over the rib cage differed when changing the impact conditions from lateral to anterior or posterior (Appendix D, see online supplement). In summary, the shoulder-to-spine coupling was reduced in the anterior impact: the scapula moved posteriorly, reducing the interaction between the subscapularis muscle and rib cage and reducing head $y$-displacement compared to the lateral impact. In the posterior impact, the amount of coupling appeared to be similar to the lateral impact: the tensed volunteers displayed similar and THUMS displayed slightly lower head $y$-displacements than in the lateral impact.

Scapula and clavicle kinematics of THUMS compared well with previous shoulder impact studies with PMHSs (Bolte et al. 2000, 2003; Compigne et al. 2004; Irwin et al. 1993; Koh et al. 2001, 2005). Compigne et al. (2004) correspondingly observed for clavicle fractures that the acromion moved more anteromedially and the scapula translated and rotated to a greater extent. They described the scapula as not being “constrained anymore by the clavicle” (p. 26). In summary, because the thorax, spine, and head kinematics did not change considerably and the scapula and clavicle kinematics in THUMS corresponded to previous PMHS experiments, the main function of the clavicle seemed to be to guide the scapula.

Though the THUMS scapula and clavicle kinematics appeared biofidelic for lateral impacts in a neutral shoulder posture, a similar assessment for elevated shoulder postures and superolateral impacts could not be conducted: this type of shoulder motion has, to the best of the authors’ knowledge, not been described in literature to date. Nonetheless, the simulation shoulder responses were expected from a biomechanical point of view. Notably, elevated shoulder postures increased the shoulder-to-spine coupling due to their preimpact scapular motion described above, resulting in larger head $y$-displacements. In superolateral impact directions, the shoulder could move more freely, reducing shoulder-to-spine coupling. In addition, the changed load direction contributed to reduced head $y$-displacements. In the parameter variation study, THUMS has been used in novel impact configurations that have not been previously reported, providing a first indication of shoulder girdle biomechanics in shoulder impacts similar to those in pedestrian crashes. Experimental validation of these findings with human subjects is recommended.

**Implications for Pedestrian Crashes**

Timing of the head kinematics is crucial to assess the effect of shoulder impacts on head injury risk in pedestrian crashes.
Head displacement away from the impacting surface, as observed in shoulder impacts, corresponds to reduced head velocity toward the vehicle in pedestrian crashes. If the time span between shoulder and head impact is sufficiently long, shoulder impacts in pedestrian crashes could help reduce the head injury risk by reducing head linear impact velocity. In contrast, if the head starts its rebound before the shoulder impact affects the head kinematics, head linear velocity caused by shoulder impact will add to the head rebound velocity, thus increasing the head delta $V$ compared to cases without a shoulder impact.

With this background, the onset of head kinematics in shoulder impacts was compared with the time span between shoulder and head impacts in pedestrian experiments. In the PMHS shoulder impact simulations, where the initial impactor speed was 4.4 m/s, it took approximately 20 ms from impactor-to-shoulder contact until the head began to move. Previous pedestrian experiment footage indicated a short time span from first shoulder contact to first head contact; for example, 6 ms in Paas et al. (2012), less than 20 ms in Kerrigan et al. (2007), and up to 30 ms in Schroeder et al. (2008), at a vehicle speed of 40 km/h. Head linear acceleration appears to be highest immediately after first head contact (Kerrigan et al. 2009). The timing implies that shoulder impact would rarely reduce skull fracture risk in the investigated shoulder impact conditions. Head impact duration was 20 ms from first contact to deepest intrusion in Paas et al. (2012), and head resultant accelerations remained high for approximately 20 ms after first contact in Kerrigan et al. (2009). For such durations (mainly observed in impacts with the windshield), head linear accelerations in the later stages of head impact may be reduced due to shoulder impact, thus reducing head injury and potentially avoiding possible head-to-dashboard contact. If head impact durations are shorter—that is, when impacting the bonnet—shoulder impacts might add to head rebound velocity, increasing the risk for traumatic brain injury. Shoulder impacts may also increase head injury risk through predeforming the bonnet or windshield prior to head impact: Predeforming the bonnet could increase head impact risk with stiff structures below the bonnet. Predeforming the windshield may damage its structural integrity, lowering its stiffness before head impact and increasing the risk of head-to-dashboard impact.

Shoulder impacts induce head lateral rotation toward the impactor. In pedestrian crashes, these rotations add to the observed head rotation toward the vehicle just before head impact, which are induced by the head catching up with the upper torso speed and rotation (Paas et al. 2012; Watanabe et al. 2011). Furthermore, the neck load around the time of head impact might be increased due to shoulder impact as a result of the induced neck curvature.

Though in-depth injury risk assessment was outside of the scope of the present study, the kinematics suggested that shoulder impacts may affect head and neck injury risk in pedestrian crashes. However, whether shoulder impacts increase or reduce the risk appears to be dependent on the exact load case; for example, head and shoulder impact location on the vehicle, bonnet-to-windshield angle, and shoulder position. Future safety systems can benefit from the results of this study. The authors believe that increasing the time span between shoulder and head impact would reduce the speed difference between the head and the vehicle and consequently reduce head injury risk.

THUMS is suitable for studying head linear kinematics and lateral rotation in shoulder impacts, similar to pedestrian crashes, for tensed subjects. THUMS head twist was considerably greater than in the volunteers; 3-dimensional head rotations must be improved for pedestrian injury prediction. The THUMS spine appeared to be stiffer than the volunteers’. Shoulder deflections in THUMS were smaller than in the relaxed volunteers but matched PMHS responses at higher severity. Overall, the THUMS response appeared to be within the expected ranges of a pedestrian of an imminent impact.

The scapular motion over the thorax had considerable influence on upper torso and head kinematics. The main function of the clavicle appeared to be to guide the scapula, whereas load transfer through the clavicle played a lesser role. Raised shoulder postures limited the scapular motion, increasing the shoulder-to-spine coupling and head linear displacements. Head linear displacements were decreased when the impact direction was changed from lateral to superolateral. It is recommended to conduct shoulder impact experiments with human subjects, similar to pedestrian shoulder impacts, to validate the results from the present study and further assess head twist. The time span between shoulder contact and head movement indicated that head kinematics during a pedestrian head impact could be altered due to shoulder impact. No general conclusion could be drawn about how head and neck injury risk are influenced by shoulder impacts; the risk is expected to depend on the exact impact conditions. This underlines the necessity to further address shoulder impacts in pedestrian experiments.

Acknowledgments

The authors thank the Japan Automobile Research Institute and the authors of Bolte et al. (2003) for providing data and support for reanalysis.

Funding

This work was funded by VINNOVA–Swedish Governmental Agency for Innovation Systems through the Fordonsstrategisk Forsknings och Innovation–Vehicle and Traffic Safety research programme and SAFER–Vehicle and Traffic Safety Centre at Chalmers, Sweden. Project partners were Autoliv Research AB, Volvo Car Corporation, the Royal Institute of Technology, and Umeå University in Sweden. The simulations were conducted on computational resources provided by the Swedish National Infrastructure for Computing at the Chalmers Centre of Computational Science and Engineering.

Supplemental Materials

Supplemental data for this article can be accessed on the publisher’s website.
References

Bolte JH, Hines MH, Herriott RG, McFadden JD, Donnelly BR. Shoulder impact response and injury due to lateral and oblique loading. *Stapp Car Crash J.* 2003;47:35–53.

Bolte JH, Hines MH, McFadden JD, Saul RA. Shoulder response characteristics and injury due to lateral glenohumeral joint impacts. *Stapp Car Crash J.* 2000;44:261–280.

Cavanaugh JM, Wallikko T, Malhotra A, Zhu Y, King AI. *Biomechanical Response and Injury Tolerance of the Pelvis in Twelve Sled Side Impacts.* Warrendale, PA: SAE; 1990.

Compigne S, Caire Y, Quesnel T, Verriest J-P. Non-injurious and injurious impact response of the human shoulder three-dimensional analysis of kinematics and determination of injury threshold. *Stapp Car Crash J.* 2004;48:89–123.

Hallaeli H, Gunal I. Normal range of scapular elevation and depression in healthy subjects. *Arch Orthop Trauma Surg.* 2002;122(2):99–101.

Irwin AL, Wallikko TJ, Cavanaugh JM, Zhu Y, King AI. Displacement responses of the shoulder and thorax in lateral sled impacts. In: *Proceedings of the Stapp Car Crash Conference,* Warrendale, PA, 1993.

Jarrett KL, Saul RA. Pedestrian injury—analysis of the PCDS field collision data. In: *Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, Windsor, ON, Canada, May 31–June 4, 1998.

Kamkar A, Irrgang JJ, Whitney SL. Nonoperative management of secondary shoulder impingement syndrome. *J Orthop Sports Phys Ther.* 1999;29(5):212–224.

Kemper A, Stitzel J, Duma S, Matsuoka F, Masuda M. Biofidelity of the SID-IIs and a modified SID-IIs upper extremity: biomechanical properties of the human humerus. In: *Proceedings of the 19th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, Atlanta, GA, 2005.

Kerrigan JR, Arregui J, Crandall JR. Assessment of pedestrian head impact dynamics in small sedan and large SUV collisions. *Int J Crashworthiness.* 2012;17(3):243–258.

Kerrigan JR, Arregui C, Crandall J. Pedestrian head impact dynamics: comparison of dummy and PMHS in small sedan and large SUV impacts. In: *Proceedings of the 21st International Technical Conference on the Enhanced Safety of Vehicles (ESV),* Stuttgart, Germany, 2009.

Kerrigan J, Kam C, Drinkwater D, et al. Kinematic comparison of the Polar J1 and PMHS in pedestrian impact tests with a sport-utility vehicle. In: *Proceedings of the International Research Council on the Biomechanics of Injury (IRCOBI)*, Prague, Czech Republic, 2005.

Kerrigan J, Murphy DB, Drinkwater DC, Kam CY, Bose D, Crandall JR. Kinematic corridors for PMHS tested in full-scale pedestrian impact tests. In: *Proceedings of the 19th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, Washington, DC, 2005.

Kerrigan JR, Crandall JR, Deng B. Pedestrian kinematic response to mid-sized vehicle impact. *Int J Veh Saf.* 2007;2(3):221–240.

Kibler BW. The role of the scapula in athletic shoulder function. *Am J Sports Med.* 1998;26:325–337.

Koh J, Cavanaugh JM, Mason MJ, et al. Shoulder injury and response due to lateral glenohumeral joint impact: an analysis of combined data. *Stapp Car Crash J.* 2005;49:291–322.

Koh SW, Cavanaugh JM, Zhu J. Injury and response of the shoulder in lateral sled tests. *Stapp Car Crash J.* 2001;45:101–141.

Lau G, Seow E, Lim ES. A review of pedestrian fatalities in Singapore from 1990 to 1994. *Ann Acad Med Singapore.* 1998;27:830–837.

Ludewig PM, Cook TM, Nawoczenski DA. Three-dimensional scapular orientation and muscle activity at selected positions of humeral elevation. *J Orthop Sports Phys Ther.* 1996;24(2):57–65.

Marieb EN, Hoehn K. *Human Anatomy and Physiology.* 8th ed. San Francisco, CA: Pearson Benjamin Cummings; 2010.

Marth DR. *Biomechanics of the Shoulder in Lateral Impact* [PhD dissertation]. Detroit, MI: Wayne State University; 2002.

Moseley BJB, Jobo FW, Pink M, Perry J, Tibone J. EMG analysis of the scapular muscles during a shoulder rehabilitation program. *Am J Sports Med.* 1992;20(2):128–134.

Ono K, Ejima S, Kaneoka K, et al. Biomechanical responses of head/neck/torso to lateral impact loading on shoulders of male and female volunteers. In: *Proceedings of the International Research Council on Biomechanics of Injury (IRCOBI),* Prague, Czech Republic, 2005.

Östh J, Brolin K, Carlson S, Wismans J, Davidsson J. The occupant response to autonomous braking: a modeling approach that accounts for active musculature. *Traffic Inj Prev.* 2012;13:265–277.

Paas R, Davidsson J, Masson C, Sander U, Brolin K, Yang JK. Pedestrian shoulder and spine kinematics in full-scale PMHS tests for human body model evaluation. In: *Proceedings of the International Research Council on the Biomechanics of Injury (IRCOBI),* Dublin, Ireland, 2012.

Paine R, Voight ML. The role of the scapula. *Int J Sports Phys Ther.* 2013;8:617–629.

Schoe der G, Fukuyama K, Yamazaki K, Kamiji K, Yasuki T. Injury mechanism of pedestrians impact test with a sport-utility vehicle and mini-van. In: *Proceedings of the International Research Council on the Biomechanics of Injury (IRCOBI),* Bern, Switzerland, 2008.

Shigeta K, Kitagawa Y, Yasuki T. Development of next generation human FE model capable of organ injury prediction. In: *Proceedings of the 21st International Technical Conference on the Enhanced Safety of Vehicles (ESV),* Stuttgart, Germany; 2009.

Society of Automotive Engineers. *J1733—Sign Convention for Vehicle Crash Testing.* Warrendale, PA: Author; 1994.

Society of Automotive Engineers. *J211—Instrumentation for Impact Test—Part I—Electronic Instrumentation.* Warrendale, PA: Author; 1995.

Subit D, Duprey S, Lau S, Guillemot H, Lessley D, Kent R. Response of the human torso to lateral and oblique constant-velocity impacts. *Ann Adv Automot Med.* 2010;54:27–40.

Thollon L. *Modélisation du Membre Thoracique dans le Cadre d’un Choc Latéral: Approche Expérimentale et Numérique* [PhD dissertation]. Marseille, France: Ecole Supérieure de Mécanique de Marseille, Université de la Méditerranée Aix-Marseille-II; 2001.

Toyota Motor Corporation. *THUMS User Manual, AM50 Pedestrian/Occupant Model, Academic Version 4.0, 2011.* Author; 2011.

Watanabe R, Katsuhara T, Miyazaki H, Kitagawa Y, Yasuki T. Research of the relationship of pedestrian injury to collision speed, car-type, impact location and pedestrian sizes using human FE model (THUMS Version 4). *Stapp Car Crash J.* 2012;56:269–321.

Watanabe R, Miyazaki H, Kitagawa Y, Yasuki T. Research of collision speed dependency of pedestrian head and chest injuries using human FE model (THUMS Version 4). In: *Proceedings of the 22nd International Technical Conference on the Enhanced Safety of Vehicles (ESV),* Washington, DC, 2011.

Weisner WM, Lee TQ, McMaster WC, McMahon PJ. Effects of simulated scapular protraction on anterior glenohumeral stability. *Am J Sports Med.* 1999;27:801–805.