Quantitative evaluation of the occupant kinematic response of the THUMS 50th-percentile male model relative to PMHS laboratory rollover tests

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ABSTRACT

Objective: The objective of the current study was to evaluate the whole-body kinematic response of the Total Human Model for Safety (THUMS) occupant model in controlled laboratory rollover tests by comparing the model response to postmortem human surrogate (PMHS) kinematic response targets published in 2014.

Methods: A computational model of the parametric vehicle buck environment was developed and the AM50 THUMS occupant model (Ver 4.01) was subjected to a pure dynamic roll at 360°/s in trailing-side front-row seating position.

A baseline configuration was defined by a baseline posture representing the average of all PMHS postures, with a friction coefficient of 0.4 for the belt and 0.6 for the seat.

To encompass challenges in controlling boundary conditions from the PMHS tests and ensure the robustness of the model evaluation, a total of 12 simulations were performed to investigate the following:

1. The effect of initial posture by adding 3 additional postures representing PMHS extremes.
2. The effect of belt tension by varying tension from the nominal vehicle retractor belt tension of 5 N to the 35 N belt tension used in the PMHS tests.
3. The effect of friction between the environment (belt, seat) and THUMS.

Trajectories (head, T1, T4, T10, L1, and sacrum), spinal segment rotations (head-to-T1, T1-to-T4, T4-to-T10, T10-to-L1, and L1-to-sacrum) relative to the rollover buck and spinal segment elongation/compression calculated from the simulations were compared to PMHS corridors using a correlation method (CORA).

Results: THUMS baseline response showed lower correlation (overall CORA score = 0.63) with the PMHS response in rollover compared to other crash modes. THUMS and PMHS demonstrated similar kinematic responses in the longitudinal axis and vertical axis but significantly different lateral excursion relative to the seat. In addition, no spinal elongation was observed for THUMS compared to the PMHS.

The posture, pretension, and belt frictions were found to alter model kinematics, especially on THUMS lateral axis motion. The posture was judged to be the most sensitive parameter evaluated because a change of 30 mm in the lateral axis results in up to an 80 mm of change in observed displacement.

Conclusions: Though the model response in the lateral axis is significantly different than that of the PMHS, it is unclear whether this difference is the result of extrinsic factors (posture, pretension, and friction), where exact values in experiment are unknown or by model intrinsic factors (e.g., spine stiffness). These differences in occupant kinematics could potentially subject the PMHS and THUMS to very different loading conditions under roof impact in rollover crashes: different occupant posture and different roof impact location. Therefore, different injury mechanisms and severity might be predicted by the current model relative to the PMHS. Consequently, though the information provided in the current study could be useful for improving model biofidelity for rollover crashes, additional studies are required to properly solve this issue.

Introduction

The number of people who die from or are injured in motor vehicle crashes (MVCs) has been declining in the past several decades (NHTSA 2014) thanks in part to the introduction of new passive safety technologies (e.g., side curtain airbag; pretensioners). However, rollover crashes still remain a serious public health problem in the United States. The proportion of rollover crash-related fatalities is increasing, from 27.9% in 1982 to 34.6% in 2012 despite the fact that rollover crashes accounted for less than 3% of the total crashes since 2010 (NHTSA 2014). Such statistics highlight the inherent need to understand occupant injury that is associated with rollover crash.

Finite element (FE) human body models offer some promising advantages as advanced injury prediction tools in injury biomechanics to quantify occupant responses, understand injury mechanisms, and develop effective injury countermeasures. Human body models (HBMs) require appropriate

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Color versions of one or more of the figures in the article can be found online at www.tandfonline.com/gcipi.

Associate Editor Joel Stitzel oversaw the review of this article.

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

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verification and evaluation of the injury prediction capability and biofidelity (e.g., global kinematics, interaction with the environment) to be considered predictive within the assumptions and limitations of the model (Rhule et al. 2002). Though head, thorax, and spine injuries are the most common severe trauma (Maximum Abbreviated Injury Scale 3+) sustained by restrained occupants in rollover crashes (Bedewi et al. 2004), their injury mechanisms are still unknown. Axial loading has been hypothesized to be the primary injury mechanism of cervical spine injury in rollover crashes (Nightingale et al. 1996) and the injury outcome and severity of the reported spine injury was dependent on the head–neck complex posture and orientation upon impact loading (Nightingale et al. 1996; Toomey et al. 2009). Because posture and orientation are dictated by occupant kinematics in rollover crashes, the ability of HBMs to predict accurate occupant kinematics in rollover crashes is vital for injury prediction.

The biofidelity of the HBM response in rollover crashes has been typically evaluated at the substructural levels (e.g., head–neck complex) under axial compressive loading (Eggers et al. 2005; Hu et al. 2008; Mato et al. 2015). To the authors’ knowledge, only one study to date has attempted to evaluate the kinematic response biofidelity of a finite element model under rollover crash conditions (Hu et al. 2009); other studies have focused on multibody model assessment (Adamec et al. 2005; Lai et al. 2005). Hu et al. (2009) evaluated the Total Human Model for Safety, or THUMS (Ver 1.52) FE model in rollover conditions, but their approach included only a comparison of measured and predicted peak head vertical excursion. In addition, the validation data used in their study were compiled from multiple existing volunteer spin tests (rollover-like) with few details on the boundary conditions (e.g., no roll time history was provided). Finally, comparing THUMS model responses with the volunteer responses as in Hu et al. (2009) might be misleading, because THUMS (Ver 1.52) was designed to mimic passive occupant responses in crashes. Benchmarking the performance of passive HBM like THUMS relative to postmortem human subjects (PMHS) tests and then adding the muscle elements to simulate the active muscle responses in rollover crashes could be an alternative.

Recently, Lessley et al. (2014) quantified the 3D whole-body kinematics of 4 PMHS restrained with a 3-point belt within a parametric rollover buck under different kinematic test conditions using a 3D motion capture technology. These data provided detailed occupant motion information and biofidelic response targets necessary for a comprehensive HBM kinematic response biofidelity evaluation.

The THUMS model can be used to evaluate detailed human response and complex injury mechanisms and injury prediction and thus is the focus of this study. Though this model has been evaluated under other impact scenarios (Poulard et al. 2014; Poulard, Kent, et al. 2015; Poulard, Subit, Donlon, and Kent 2015), its performance under rollover conditions has not been assessed. As part of a larger effort to obtain a biofidelic HBM for occupant response simulation in rollover crashes, the goal of this study is to evaluate THUMS kinematic response biofidelity against the PMHS response corridors presented by Lessley et al. (2014).

Methods
A computational model of the parametric vehicle buck environment was developed and the AM50 THUMS occupant model (Ver 4.01; Shigeta et al. 2009) was subjected to a pure dynamic roll at 360°/s.

A baseline configuration was defined including a baseline posture representing the average of all PMHS postures and a friction coefficient of 0.4 for the belt and 0.6 for the seat.

To encompass challenges in controlling boundary conditions in the PMHS tests and ensure the robustness of the model evaluation, a total of 12 simulations were performed to investigate the following:

1. The effect of initial posture by adding 3 additional postures representing PMHS extreme postures.
2. The effect of belt tension by varying tension from the nominal vehicle retractor belt tension of 5 N to the 35 N belt tension used in the PMHS tests.
3. The effect of friction between the environment (belt, seat) and THUMS.

Overview of the experiments
THUMS response was evaluated relative to the PMHS response corridors presented by Lessley et al. (2014). The PMHS tests were performed with the Dynamic Rollover Testing System (DRoTS), which was designed to perform controlled and repeatable rollover crash tests in the laboratory (Figure 1a). The DRoTS was used to drop/catch and rotate a rollover test buck around its center of gravity (CG) without roof-to-ground contact. The rollover test buck replicated the occupant compartment of a typical mid-sized SUV developed for the U.S. market (Foltz et al. 2011; Toczyski et al. 2013) and was designed to approximate a vehicle-based restraint environment with simplified boundary conditions (constructed mostly of steel), which allowed ease of test reproducibility and the facilitation of computational modeling (Figure 1b). Using a 3D optic motion tracking system, the PMHS whole-body 3D kinematic response corridors at 9 different test conditions were constructed, including 2 seating locations (trailing side vs leading side), 2 belt pretension levels (35 and 300 N), and 5 different buck CG kinematic conditions (Lessley et al. 2014).

The PMHS tests that were used to evaluate the HBM in this study are the PR-T tests as described in Lessley et al. (2014). These tests are a pure dynamic roll test with the occupant seated in the trailing side position (Figure 1b) and 3-point belted with 35 N belt pretension force. This test condition was selected because the occupant kinematics in these tests (PRT-T) are characteristic and representative across all kinematic conditions.

Rollover simulation with THUMS
The finite element rollover buck model was developed for LS-DYNA solver based on 3D CAD drawings (Figure 1c). A number of key components were modeled as rigid bodies (main frame, base plate, seat plate, back support plate, D-ring, outboard anchor, inboard anchor, rod, buckle, foot plate,
and knee bolster). For better decomposition results in MPP LS-DYNA, the model only retained necessary parts for the occupant–buck interaction. The simplified buck model has 254,091 nodes and 221,797 elements (128,221 shell elements; 93,576 solid elements).

The HBM was then seated and positioned in the buck model as it was in the tests. Presimulations were run to adjust its posture and settle the occupant down by applying gravity (Appendix A, online supplement). Strains and stresses obtained after settling down were accounted for during the rolling simulation. The HBM was then 3-point belted and a pretension force of 35 N was applied at both of the outboard ends of the shoulder and lap belts. This was achieved by applying prescribed displacement to the outboard ends of the shoulder and lap belts before initiating the rolling motion.

During the preparation of the PMHS tests conducted in Lessley et al. (2014), the head of the PMHS was held up by tape in order to maintain the head. The tape was cut so that it was supposed to break at the beginning of the roll. However, it was hypothesized that the tape did not break initially and in fact may have actually held the head up into the roll during the first 200 ms (12° of roll). In order to replicate its effect on the head, cable elements were attached between the head (forehead and chin) and the buck. The cut was not modeled but directly included in the material parameters of the cable, which was determined from experiments ($E = 0.4$ GPa, $\sigma_p = 40$ MPa, $\varepsilon = 1\%$).

Finally, the CG roll rate time history (Figure 1d) measured from the test was applied to the rollover buck to simulate the buck rotational motion. The pure dynamic roll event lasts 1500 ms with a peak angular rate of $360^\circ$/s. The friction coefficients in the simulations were defined according to the THUMS model frontal impact sled test simulations (Poulard, Kent, et al. 2015) as follows: (1) 0.4 between the occupant and seat belt; (2) 0.4 between the seat belt and seat belt slip ring; (3) 0.6 between the occupant and buck structure.

In the PMHS rollover tests, a single kinematic measurement marker was placed on each of the measured spinal vertebrae (T1, T4, T10, L1, and sacrum) and shoulders (left and right shoulder acromion) for their 3D kinematics and a 4-marker cluster was placed on the skull for the head CG 3D kinematics (Lessley et al. 2014). In addition, the PMHS spine was divided into 5 segments (head to T1, T1 to T4, T4 to T10, T10 to L1, and L1 to sacrum) to describe additional kinematics of the spine, including segment rotations (along the local X-axis and Y-axis) as well as potential spinal elongation and compression.

In the current study, the markers were attached to the HBM in the simulation at the same spinal vertebrae and shoulder acromia as in the experiment for the output of the THUMS spinal 3D kinematic response. A total of 42 kinematic responses were calculated from the occupant kinematic data for each simulation, which included (1) 3D trajectories at head, left and right acromia, T1, T4, T10, L1, and sacrum; (2) spine segment rotations (x-axis, y-axis) relative to the buck (SAE vehicle coordinate system) at head-to-T1, T1-to-T4, T4-to-T10, T10-to-L1, L1-to-sacrum, and spine segment deformation (Figure B1 in Appendix B, online supplement).

Quantitative assessment of THUMS kinematic response biofidelity

To quantitatively assess the biofidelity of the kinematic response of the HBM, the correlation and analysis (CORA) method was used in this study (Gehre et al. 2009). This method consists of a cross-correlation method (magnitude coefficient, shape coefficient, phase coefficient) and a corridor method. For the corridor method, the inner and outer corridors were defined to be the average experimental response plus or minus $\alpha$ standard deviation (SD), where $\alpha = 1$ for the inner corridor and $\alpha = 2$ for the outer corridor. The CORA rating score is calculated, by assigning a weighting factor to each of the subscores (0.2 for the magnitude, 0.2 for the shape, 0.2 for the phase, and 0.4 for the corridor) as recommended in ISO/TS 18571 (Barbat et al. 2013). The CORA score was calculated for 3D displacements (head, T1, T4, T10, L1, and sacrum), spinal segment (head-to-T1, T1-to-T4, T4-to-T10, T10-to-L1, and L1-to-sacrum) rotations relative to the rollover buck, and spinal segment extension/compression and spinal intersegment rotations.

Simulation parameter sensitivity study

Though extensive care was taken for a well-controlled test protocol and to facilitate computational modeling of the PMHS tests,
some uncertainties of the experimental condition still exist. Therefore, to ensure robustness of the model validation, the sensitivities of the HBM kinematic responses to the parameters with uncertainty in the experiment were assessed in this study.

A matrix of 12 simulations (Table 1) was performed. This included sensitivities to the seat belt pretension force, the friction coefficient between THUMS skin and the buck interior, and the occupant seating posture. The belt tension was evaluated at 35 N (baseline: the target pretension force in experiment) and at a nominal vehicle retractor belt tension of 5 N. The friction coefficient for the belt and seat was evaluated independently at 3 levels for the belt and the seat; belt vs. skin: 0.2, 0.4 (baseline), 0.6; seat vs. skin: 0.4, 0.6 (baseline), 0.8.

Finally, the effect of occupant seating posture was evaluated. Though the PMHS in the rollover tests was positioned in a manner that closely approximated the University of Michigan Transportation Research Institute standard driving position (Lessley et al. 2014), variability always exists to some extent, due to the variability in occupant sizes tested and the occupant posture/positioning measurement technique. In total, 4 different occupant seating postures were evaluated with the HBM: A, B, C, and D (Figure 2). Posture A was based on the average PMHS positioning measurements from the tests presented by Lessley et al. (2014) and defined as the baseline posture. Key measurements included the H-point location, which was defined as the distance between the greater trochanter of the femur bone and the lower block of the rollover buck seat back. Posture B posture deviated from the baseline posture with the H-point move forward by 20 mm and a 5° pitch rotation of the HBM head–neck–torso due to settling. Posture C deviated from posture A with the H-point move laterally by 30 mm closer to the center in order to modify the belt path (belt away from the neck). Finally, posture D is a combination of postures B and D.

### Table 1. Simulation matrix.

| # | Posture | Belt tension (N) | Belt friction | Seat friction |
|---|---------|------------------|--------------|--------------|
| 1 | A       | 35               | 0.4          | 0.6          |
| 2 | B       | 35               | 0.4          | 0.6          |
| 3 | C       | 35               | 0.4          | 0.6          |
| 4 | D       | 35               | 0.4          | 0.6          |
| 5 | A       | 35               | 0.2          | 0.6          |
| 6 | A       | 35               | 0.6          | 0.6          |
| 7 | A       | 35               | 0.4          | 0.4          |
| 8 | A       | 35               | 0.4          | 0.8          |
| 9 | A       | 5                | 0.4          | 0.6          |
| 10| B       | 5                | 0.4          | 0.6          |
| 11| C       | 5                | 0.4          | 0.6          |
| 12| D       | 5                | 0.4          | 0.6          |

### Solver and hardware configuration

LS-DYNA (LSTC, Livermore, CA) version R7.1.1, Distributed Memory Parallel (mpp) version with explicit solver was used to perform all rollover simulations. All simulations were performed using the University of Virginia’s state-of-the-art Rivanna HPC computational cluster, which is a ∼5,000 core (Intel Xeon E5-2670v2, 2.5 GHz, 20 core) university-wide resource. In order to maximize total run time and eliminate decomposition performance penalty across multiple nodes, all jobs were run using a single computational node for each simulation.

### Results

A total of 12 simulations were performed with THUMS. All simulations terminated normally at 1,500 ms. All of the 42 kinematic responses that characterize the human motion relative to buck were calculated for each of the simulations (Appendixes C to G, online supplement). The data are plotted relative to the buck roll angle to facilitate understanding the occupant response data. Figure 3 provides snapshots for every
Figure 3. Rollover sequence for the baseline simulation (simulation #1).

45° angle or rotation of simulation #1, which is the baseline simulation.

During the tests, THUMS demonstrated 3D motions relative to the buck, in a manner similar to that of the PMHS. The buck rotational motion imposed centrifugal accelerations to the occupants, directed radially from the buck rotational axis to the buck exterior. The centrifugal accelerations pulled THUMS both outboard (along the buck local Y-axis) and upward (along the buck local Z-axis). In addition, this upward and outward motion was accompanied by the THUMS lateral rotation along the buck local X-axis in the coronal plane and the occupant pitch rotation along the buck local Y-axis in the sagittal plane. This is because THUMS tended to straighten out and extend its spine to align with the resultant acceleration vector as the PMHS did, due to the compliance in its spine. Lastly, this pitch motion resulted in the THUMS occupant translational motion (along the buck local X-axis).

However, some response discrepancies were observed (Appendix C). First, THUMS upper body (head, T1, and T4) moved up by 40 mm during the first 15° of the roll (corresponding to 150 ms), whereas the PMHS did not exhibit this behavior (Appendixes Cb and Cc). This is possibly due to (1) the inertia keeping THUMS in place and (2) the stiffness of the tape which was used for securing the head. The head of THUMS showed an oscillating response not observed in the PMHS tests. This oscillating response carried away the head in all directions up to 70 mm from where the PMHS head was located. This leads up to 110 mm of deviation for the head impact location on a possible touchdown (reported between 100° and 300°).

In addition, the lower body (T10, L1, sacrum) of the HBM moved outboard (Y-axis) more than the lower body of the PMHS (Appendixes Cd, Ce, and Cf). The left shoulder acromion of THUMS moved rearward and stabilized around 100° or roll, whereas the PMHS left shoulder acromion stabilized after 150°, suggesting higher coupling in the model than in the PMHS. The HBM right shoulder acromion displayed higher lateral excursion in the early stage of the rollover, and in the later stages of rollover (roll angle > 100°) it did not reproduce the change in direction observed in the PMHS (Appendixes Cg and Ch). This resulted in a larger shoulder segment axial rotation (along the Z-axis) in the HBM simulation than in the PMHS tests (Appendix Ci).

Finally, the lumbar and thoracic spine kinematics showed some discrepancies. No elongation of the HBM spine was observed, contrary to the PMHS where the spine elongated or compressed (Appendix Ck). Then, THUMS experienced high segment bending about the X-axis around the L1 vertebrae (Appendixes Cl and Cm), even showing contrary rotation (Appendix Cm). Larger discrepancies in spine bending were observed around the Y-axis (flexion/extension) for T4 vertebrae where the THUMS spine exhibited more extension (Appendix Cn) than L1 vertebrae where the THUMS spine exhibited more flexion (Appendix Co).

Quantitatively, the THUMS kinematic responses achieved an average CORA rating score of 0.60 ± 0.23 for the baseline simulation (simulation #1 of Table H1 in Appendix H, online supplement). The CORA score obtained for the 3D displacements (0.72 ± 0.15) was higher than the score obtained for the rotations (0.52 ± 0.24) and the elongations (0.26 ± 0.12). The head yielded a better CORA rating (0.71 ± 0.19) than the other body regions and L1 obtained the lowest score (0.45 ± 0.23).
Figure 4 shows the difference in spine orientations at 180° (touchdown) due to seat belt friction (a), seat friction (b), posture (c) and belt pretension (d).

A comparison of all kinematic measurements among simulations with various friction coefficients is presented in Appendix D (seat belt friction) and Appendix E (seat friction). The seatbelt friction affected model response by affecting the forward and vertical motion of the head and T1 even its effect was reduced at the end of the rollover cycle (360°). With a seat belt coefficient of 0.8, the head was further away (20 mm) from the PMHS corridor at the end of the cycle, suggesting a less realistic value for friction. Similar trends were observed for the seat friction (Appendix E), with these effects especially visible for the sacrum. At 180° of roll angle, the spine orientation did not differ due to friction (Figures 4a and 4b). Though CORA scores for specific body regions were affected by friction changes (upper body for seat belt friction, sacrum for seat friction), CORA scores were not improved.

A comparison of all kinematic measurements among simulations with various postures is presented in Appendix F. With a more forward posture by 20 mm (posture B), it results in a 40 mm change in observed displacement for the head (rearward). With the HBM moved laterally by 30 mm (posture C), it results in up to an 80 mm change in the lateral displacement (outward) for the overall body. Though posture D (B + C) is similar to posture C in terms of HBM motion for the lateral axis, the HBM motion is similar to posture B in the antero-posterior axis (forward). Finally, occupant initial posture affected directly the spinal segment rotation of the upper body (y-axis) and the lower extremity (x-axis). At 180° of roll angle, the spine orientation differed significantly due to posture (Figure 4c).

A comparison of all kinematic measurements between simulations with 35 N seat belt pretension (baseline) and 5 N is presented in Appendix G. With lower pretension force (5 N), the model exhibited larger displacement in all 3 directions due to the additional slack induced in the seat belt restraint system. At 180° of roll angle, the spine orientation variability due to belt pretension was not observable (Figure 4d).

**Simulation parameter sensitivity study**

Despite the intensive efforts toward the development of HBMs (Yang et al. 2006), studies that evaluated the biofidelity of their whole-body 3D kinematic response are rare. This is primarily because such detailed 3D occupant kinematic response data were only available recently with the use of the video-based optoelectronic stereo-photogrammetry methods like the method presented by Lessley et al. (2011). This study is the first attempt to assess the whole-body 3D kinematic response biofidelity of a human body FE model under rollover crash condition. Specifically, the kinematic response of THUMS was assessed relative to the PMHS response corridors presented by Lessley et al. (2014). This test condition is characteristic and representative, because the PMHS in this condition in general demonstrated similar 3D motions relative to the buck as the PMHS in other conditions (Lessley et al. 2014). Furthermore, the pure dynamic rollover condition was selected because it was inherently the most controlled test type because it did not also include a superimposed vertical acceleration. Finally, we focused on the occupant in the trailing-side position, because the occupant in the trailing-side seating position has been reported to have a higher risk of injury than the occupant in the leading-side (Funk et al. 2012; Parent et al. 2011; Viano et al. 2007).

The model was able to reproduce the predominant occupant motions observed in the PMHS: drifting outboard and upward during the tests similar to the PMHS as the ATD previously evaluated (Zhang et al. 2014) but, contrary to ATD, the model head and upper torso pitched rearward as the PMHS due to the flexion/extension compliance in the model spine. In addition, the model showed some response discrepancies compared to the PMHS response. The discrepancies of the spine lateral displacement and spinal intersegment rotation/bending between the model and the PMHS suggested potential differences in the trunk stiffness (combination of spine, ribcage, and soft.
tissues/organs) between THUMS and the PMHS. These differences in head–neck complex posture and orientation lead up to the 110 mm of deviation for the head impact location observed between THUMS and PMHS at 180° of roll. These differences in head–neck complex conditions under roof impact in rollover crashes could lead to potential different injury mechanisms (Nightingale et al. 1996; Toomey et al. 2009).

It was hypothesized that the neck muscles of THUMS Ver 4.0 may be stiffer than the average response of a PMHS in lateral impact (Poulard et al. 2014). The THUMS Ver 4.0 neck was evaluated under dynamic axial loading (Nightingale et al. 1997), but no comparison with the experimental was reported (Toyota Motor Corporation 2011). Additional evaluation of the spine of THUMS Ver 4.01 at a hierarchical approach from the component level (local) to the structure level (global) would help determine the correct model responses. Such component-level assessment has the advantage of a more simplified and controlled loading condition compared to rollover where boundary conditions uncertainty will affect model responses. It would also be useful to assess the intervertebral disc and spine ligament response in order to investigate the lack of spine elongation/compression observed for the HBM. Relevant evaluation of the HBM for the rollover crash condition could include evaluating the functional spine unit level (Lopez-Valdes et al. 2014; Markolf 1972; Panjabi et al. 1976) and whole lumbar spine level (Begeman et al. 1994; Demetrooulos et al. 1998).

A response sensitivity analysis was also conducted in this study to encompass challenges in controlling boundary conditions in the PMHS tests and ensure the robustness of the model evaluation. The sensitivity analysis in this study indicates that the model responses depend on factors such as the belt pretension, friction coefficient, and posture of the occupant. Furthermore, due to the nature of omnidirectional loading conditions in rollover crashes, the direct effect of such factors on the HBM kinematic response in one specific direction/axis could also lead to a change in kinematic response in other directions/axes. This highlights the challenge of evaluating HBM responses and their use to simulate occupant responses in rollover conditions. From all of the different parameters evaluated, the posture was judged to be the most sensitive parameter evaluated because a change of 30 mm in the lateral axis results in up to 80 mm of change in observed displacement. This sensitivity to posture suggests that further efforts should be allocated to a better quantification of the posture in further experiments. Although the study presented by Lessley et al. (2014) represents the most detailed measurement of PMHS kinematic response under rollover crashes to date, a more comprehensive characterization of spinal kinematics will be helpful for a better assessment of the model kinematic response. Specifically, including the 4-target cluster technique developed by Lessley et al. (2011) will permit the exact quantification of the initial occupant posture (vertebrae CG position and orientation) and its implementation in THUMS (Poulard, Subit, Donlon, and Kent 2015). Therefore, a personalized approach could be used to study the effects of preimpact posture on occupant responses, as done previously in frontal impact (Poulard, Subit, Nie et al. 2015) and side impact condition (Poulard et al. 2014). The authors recommend that model users introduce variability in head–neck initial posture during rollover crashes with Original Equipment Manufacturer (OEM) vehicles to increase the range of head impact conditions.

This study aims to evaluate the passive response of the THUMS model (no active muscle responses) relative to the PMHS response. Future study should also improve and validate the active muscle tensing responses of the THUMS model relative to volunteer tests. This would also require modeling of the active muscle response with a closed-loop feedback control system, in order to simulate such a long-duration event.

Furthermore, even the rollover condition used in this study is a simplification of real-world rollover crashes. This study only simulated a portion of the rollover crashes—the free-flight rotation phase. The vehicle roll initiation phase before the free-flight rotation phase is not simulated. This is important to consider because such a roll initiation phase could bring the occupant out of position. Therefore, the occupant posture at the beginning of the free-flight rotation phase of a real-world rollover crash could be different than the nominal posture used in the current study.

Therefore, future studies should also consider occupant kinematics in other phases of rollover crashes.

In addition, even the simulation of free-flight rotation phase in this study has limitations. The PMHS experiments were conducted using a ground-based fixture to study occupant kinematics. This is probably because the use of a ground-based fixture provided a practical way for testing occupant responses under the free-flight rotation phase. However, this type of test is different from the free-flight rotation phase in a real rollover crash, where gravity does not affect the occupant motion relative to the vehicle because the effects of gravity are canceled out. This is because both the vehicle and occupant are falling under gravity in a real rollover. Therefore, this means that when analyzing occupant kinematics relative to the vehicle in ground-based fixture test, the effect of gravity needs to be taken into account.

The goal of this article is not to simulate the occupant responses in rollover as realistic as possible. Rather, this article should be considered one step toward the ultimate goal of developing a biofidelic computational human body models for simulating (living) occupant responses in real rollover crashes.

The kinematic passive response of THUMS was evaluated relative to the PMHS kinematic response in laboratory rollover tests. To ensure a more robust model response validation/assessment, a sensitivity analysis of the model responses to parameters with uncertainties in the experiment was also conducted, including seat belt pretension, friction coefficient, and occupant seating posture. The model was able to reproduce the predominant occupant motions observed in the PMHS: drifting outboard and upward and head and upper torso pitching rearward due to the flexion/extension compliance in the model spine. In addition, response discrepancies between the model and the PMHS were observed in terms of the thoracic/lumbar spine and shoulder kinematics, suggesting that modeling efforts should be applied to the spine.

Though all of the parameters affected model kinematics, especially lateral axis motion, the posture was judged to be the most sensitive parameter. Though some posture change targeted a specific direction/axis, a potentially higher change in kinematic response in other directions/axes was observed. This
highlights the challenge of evaluating HBM responses and their use to simulate occupant responses in rollover conditions.

This study is the first attempt to evaluate the whole body 3D kinematic response biofidelity of a human body FE model in a controlled rollover and should be considered one step toward the ultimate goal of developing a biofidelic computational human body model for simulating (living) occupant responses in real rollover crashes.

Funding

This work was supported by funds from the Toyota Collaborative Safety Research Center (Toyota CSRC). The views expressed in this article are those of the authors and do not necessarily represent or reflect the views of the sponsors.

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