Comparison of Human Body Model and PMHS Occupant Kinematic Response in Laboratory Rollover Tests

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ABSTRACT

To date, no human body computational models (HBM) have been shown to possess any level of biofidelity in predicting human occupant responses in rollover crashes. The goal of this study was to assess the kinematic response biofidelity of a HBM for rollover crash simulations by comparing model kinematics to biofidelity targets from the literature. The TASS MADYMO Active Human Model was chosen as the HBM for this study because it is hypothesized that the active musculature will have a greater effect on kinematic response in rollover crashes compared to other crash modes. Considering the range of possible occupant responses—from atonic to fully tensed—this study compared the passive response of the HBM, which was examined by disabling the active features, to post mortem human surrogate (PMHS) response in controlled laboratory rollover tests from the literature which consisted of four PMHS subjected to a passenger-side leading pure dynamic roll (360 deg/s). The kinematic trajectories was generated from tracking reflective markers attached to the head, T1, T4, T10, L1, sacrum, and left/right acromion. In the current study, the HBM was seated in a model of the test buck and was subjected to kinematics replicate of the test conditions. To make a comparison between the HBM and the PMHS, some assumptions had to be made due to variability in PMHS anthropometry, restraint fit, and orientation with respect to seat position. Thus, to understand the implications of the various assumptions, a sensitivity analysis was conducted that varied seat friction, pre-test restraint tension, and occupant positioning. After making some minor modifications to the HBM, the overall response predicted by the HBM fell generally within the published corridors. Specifically all of the upper body kinematic trajectories remained within the YZ (lateral) trajectory corridors for the PMHS test conditions. However, as indicated to be a problem with crash dummy biofidelity, the fore-aft (XZ) motion of PMHS was not accurately predicted by the HBM but was closer in the roll range in which a possible impact would occur. The imposed (vehicle) kinematics and restraint resulted in a low sensitivity of peak lateral and vertical excursion to initial position. However, varying the belt tension resulted in relatively large changes in kinematic response. Now that the model has been shown to have satisfactory kinematic biofidelity in the atonic configuration, it will be used to examine the effect of variations in active muscle tensing to lend insight into protecting humans in rollover crashes.

INTRODUCTION

According to the National Highway Traffic Safety Administration (NHTSA), there were 33,561 casualties and 2.3 million people injured in automobile accidents in 2014. While rollover events only accounted for 2.2% of total crashes, they accounted for 34.6% of total fatalities. These crashes are estimated to cost upwards of 50 billion dollars annually. (NHTSA 2014) There is a clear incentive to try to understand the injury mechanisms that occur in rollover crash events so as to better design countermeasures. Several biomechanical studies (Nightingale et al., 1991; Pintar et al., 1995; Nightingale et al., 1996; Toomey et al., 2009) have demonstrated that the injury outcome and severity of the cervical spine under primarily axial loading, which has been hypothesized to be the primary injury mechanism in rollover, are highly dependent on the occupant posture and orientation during impact loading. Therefore occupant kinematics is essential in understanding the interactions that take place inside a vehicle in a rollover.

In order to understand occupant kinematics, a biofidelic tool is needed to assess what is occurring during a rollover event. The best possible tool that could be used to represent an occupant within a rollover would be a volunteer. However, volunteers can only be used in low severity cases for obvious ethical reasons. Post Mortem Human Subjects (PMHS) can be used over the whole range of severity cases. Lessley et al. (2014) conducted a series of laboratory rollover tests with PMHS using the dynamic rollover test system (DRoTS). Zhang et al. (2014) in a concurrent paper to Lessley et al. (2014) conducted tests with six of the existing Anthropometric Testing Devices (ATDs) under similar conditions. The results showed that the dummies did not match the response of the PMHS.

In addition to crash test dummies, computational models have also been frequently used for predicting occupant responses in vehicle crashes. To date, there is no study that evaluated the response of human body models in rollover events. Considering rollover crashes last longer than other crash modes (seconds versus hundreds of milliseconds), it poses some significant challenges for computational modeling: 1) high computational cost; 2) potentially significant effect of muscle tensing on occupant responses. Therefore, the MADYMO Active Human Model, created by TNO and distributed by TASS, provided a computational efficient and promising tool for studying occupant kinematics responses in rollover crashes and the effect of musculature tensing.

The purpose of this study is to evaluate the baseline response of the MADYMO Active Human Model by comparing it to the kinematics of the PMHS tests conducted by Lessley et al. (2014) using the same conditions. In order to evaluate the baseline response of the Active Human Model, the active musculature will be turned off.

METHODS

In Lessley et al. (2014), there were 36 total tests made up of 9 different test conditions conducted with 4 PMHS. DRoTS, located at the Center for Applied Biomechanics at the University of Virginia, was created in order to produce controlled, repeatable, rollover crash tests for full sized vehicles in a laboratory setting. These tests were performed using a parametric buck that is representative of the geometries and inertial properties of 12 full-sized sport utility vehicles (SUV) on the United States market (Foltz et al. 2011). For this study, the comparison between the PMHS and Active Human Model will only be of the pure roll with the occupant on the leading side of the buck. The leading side is the side that leads in the roll, in this case it is the passenger seat.

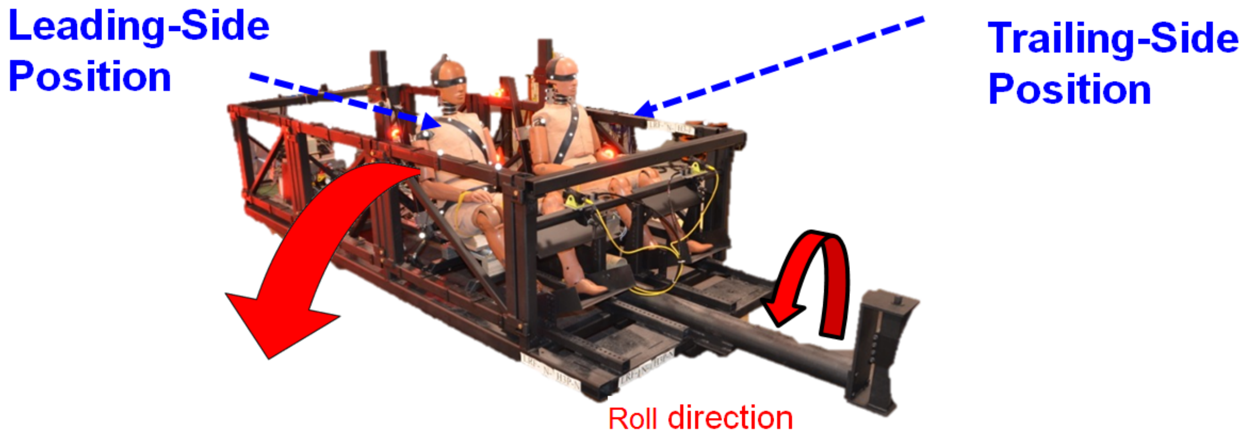

Figure 1: Parametric Buck used in Laboratory Rollover Tests

Using a CAD model of the parametric buck, a shortened mesh buck was created that was made up of shell elements composed of null material. This mesh then connected to a reference space by a revolute joint within MADYMO. A prescribed motion was imposed about the joint that was directly taken from angular rate sensors on the buck during the Lessley et al. (2014) PMHS tests.

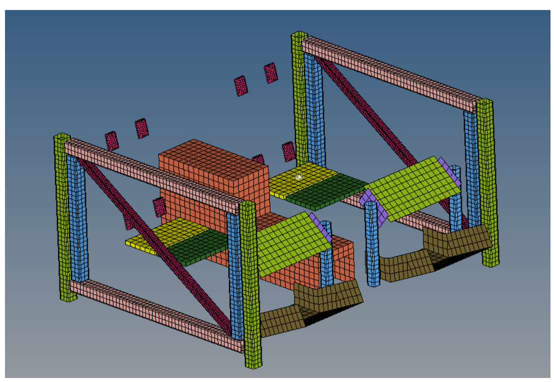

Figure 2: Meshes Buck Model

The Active Human Model was the placed on the leading side of the buck and was positioned in a similar fashion to how the PMHS were positioned. This was determined using measurement data taken of the PMHS prior to testing. The initial friction coefficient between the occupant and the buck was set at 0.4.

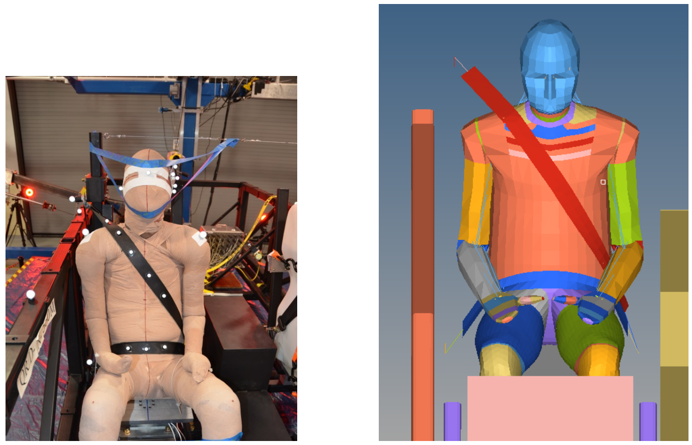

Figure 3: Initial Position of PMHS and Active Human Model

The elastic contact characteristics between the Active Human Model and the buck were determined by the characteristics incorporated within the Active Human Model. The meshed buck was considered a rigid body. The elastic deformation was modeled by the amount the slave nodes penetrated the master surface segments. The model was locked and settled with a pre-simulation. The final position of the pre-simulation was then imported as the beginning position of the main simulation.

The belt that was used was made up of shell membrane elements for contact with the model. The shell elements of the belt were connected to attachment points by 1D elements. The belt was routed on the model using measurements taken from the PMHS tests prior to the roll. The lap belt and the shoulder belt during the PMHS tests were connected by a D-ring. This was incorporated within the multibody model with friction neglected. The belts in the PMHS tests did not have a pre-tensioner or retractor. However, 35 Newtons of tension was preloaded on the belt prior to the roll by manually tightening the belts. This was represented in the MADYMO model by attaching force actuators to the belt and applying a 35 N force until the actuators reached equilibrium. The actuators were then locked into place.

In Lessley et al. (2014), the kinematics of the head cg, T1, T4, L1, Sacrum, and the right/left shoulder were analyzed using a VICON system to track the total net displacements in three dimensional space. Output markers were placed on the Active Human Model that then outputted similar kinematic data.

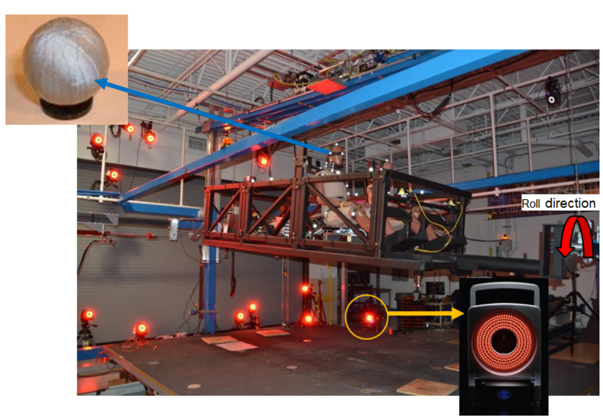

Figure 4: VICON System Setup

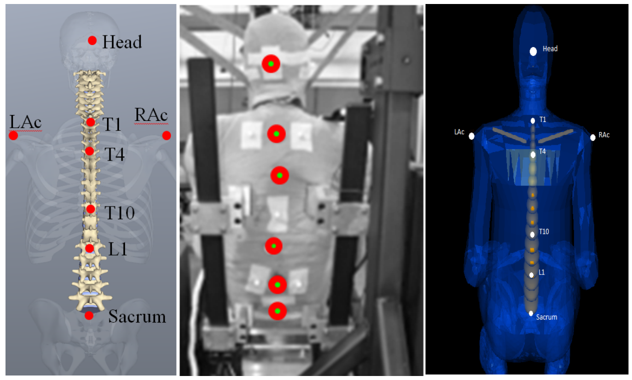

Figure 5: Output Markers for PMHS and Active Human Model

In Lessley et al. (2014), kinematic corridors were created from the displacement of the markers over the roll angle of the buck. All eight leading side trajectories were averaged together and enclosed by a standard deviation from the average.

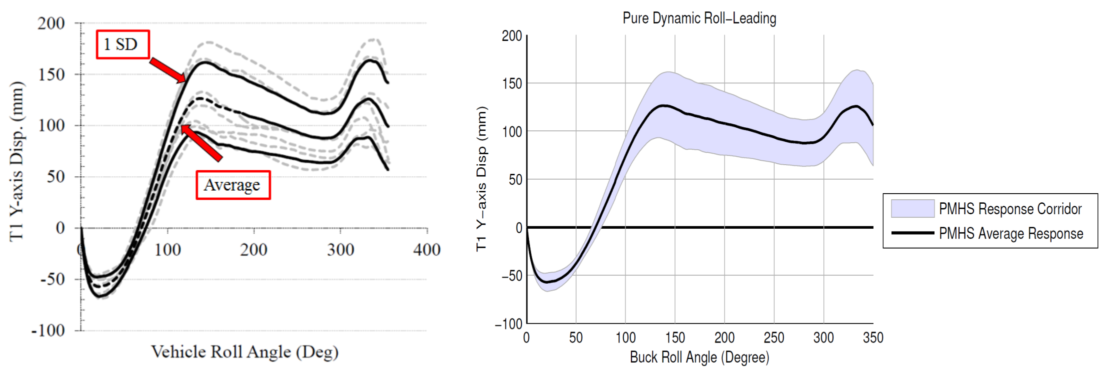

Figure 6: Eight PMHS Tests turned into Kinematic Corridor

An initial simulation was conducted and from the output markers on the Active Human Model, the corresponding trajectories were plotted over that of the PMHS corridors. Additional to the displacements, corridors for extension and rotation between markers were also calculated. All of the kinematic information is stated in the SAE vehicle coordinate system that is fixed to the buck. So as the buck rotates the coordinate system also rotated.

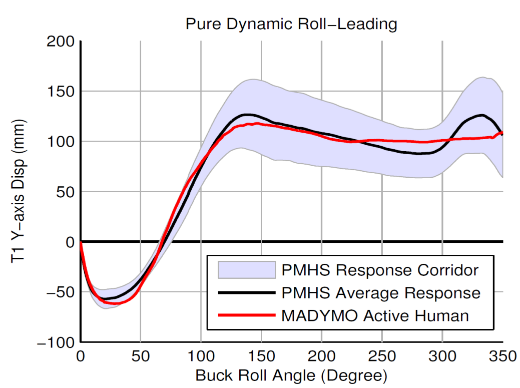

Figure 7: MADYMO Active Human Model Overlaid on PMHS Kinematic Corridor

Modelling Tape on the Leading Side

To address the difference seen in the model from the PMHS, measures were taken to try to make the Active Human Models conditions closer to that of the PMHS tests. In the PMHS tests conducted at UVA, the head of the PMHS was held up by tape. The tape was cut so as to break at the beginning of the roll. However, when looking at the video it was hypothesized that the tape did not break initially and in fact may have actually held the head up in the midst of the roll.

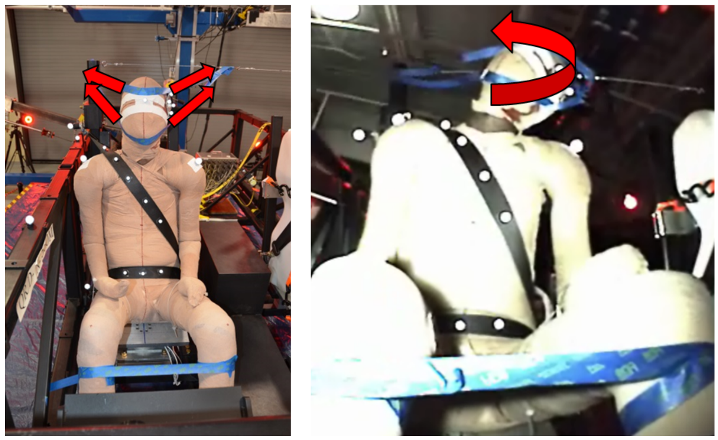

Figure 8: Tape on the Head

In order to replicate the effect on the head, belt elements were attached to nodes of the facet surface on the Active Human Model's forehead and the chin. The belt elements are 1-dimensional elements that can be specified to rupture at a certain strain. The strain at which the belt element would break was determined by watching the video of the PMHS test and finding the time in which the tape would break in the roll. The break strain was then adjusted until it broke at the correct corresponding time. This strain was determined to be 1% strain.

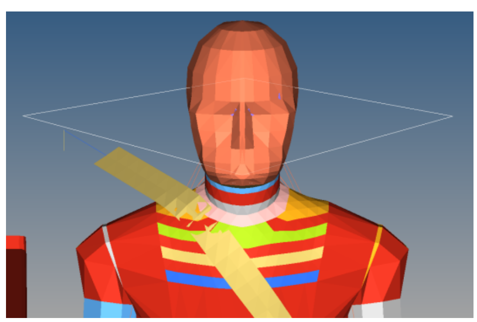

Figure 9: 1-D Belt Elements

Sensitivity to Different Boundary Conditions

Changing the Initial Position

To evaluate how the initial position would affect the occupant kinematics, five different positions were set up. The first is the baseline that was created after adding belt elements to represent the tape. The second has the arms more forward from the baseline. The third has the head neck more upright. The fourth has the thorax more upright with the pelvis touching the back seat block. The fifth and final posture has the thorax more reclined so that the pelvis is further from the back pelvis block.

Changing the Seat Friction

The seat friction coefficient for the baseline simulation was 0.4. Simulations were conducted ranging from different friction coefficients of 0.1 to 1.

Changing the Seat Belt Tension

The seat belt tension was evaluated by changing the force at which the actuators pulled on the belt. The tension in the belt was varied in range of 0 to 100 N.

RESULTS

The initial corridors were created of the leading side roll. The results of which were that the upper body of the Active Human Model in the outboard and vertical direction (Directions Y and Z) were very consistent with the PMHS corridors. As seen in Figure 10 and Figure 11.

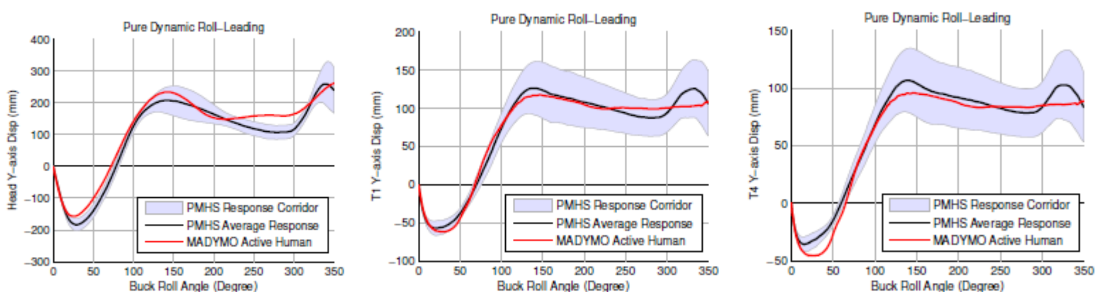

Figure 10: Upper Body Corridors in the Y Direction

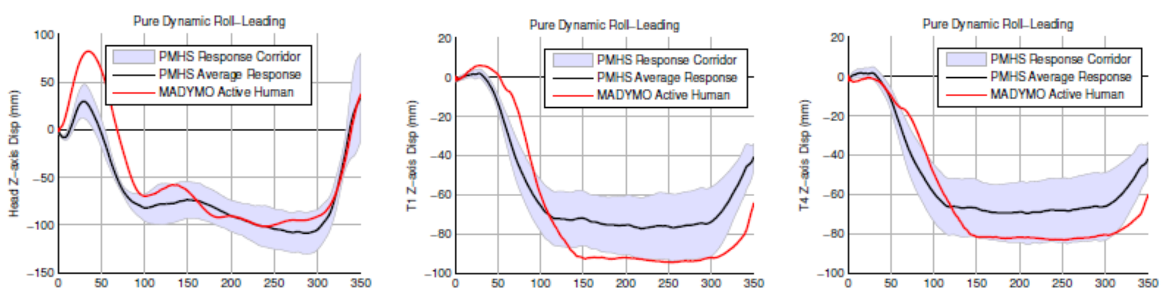

Figure 11: Upper Body Corridors in the Z Direction

However, when looking at the lower body in the Y direction, the lower the marker is on the body, the less the corridors match up. The way that the sacrum maintains its distance after

125 degrees raised the suspicion that this was a boundary condition problem with the pelvis interaction.

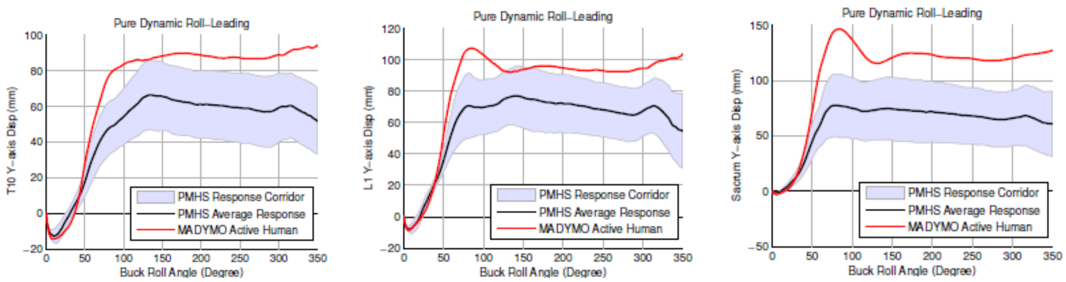

Figure 12: Lower Body Corridors in the Y Direction

Visual Comparison of the Leading Side

This similarity in the outboard and vertical directions is evident throughout the simulation.

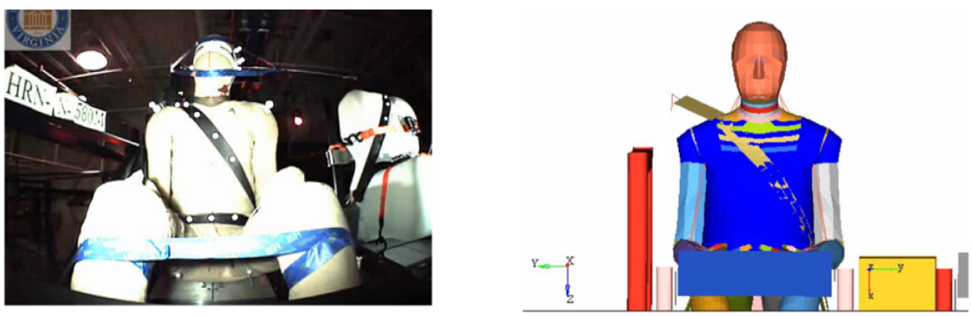
Figure 13: Front Comparison t=0 s

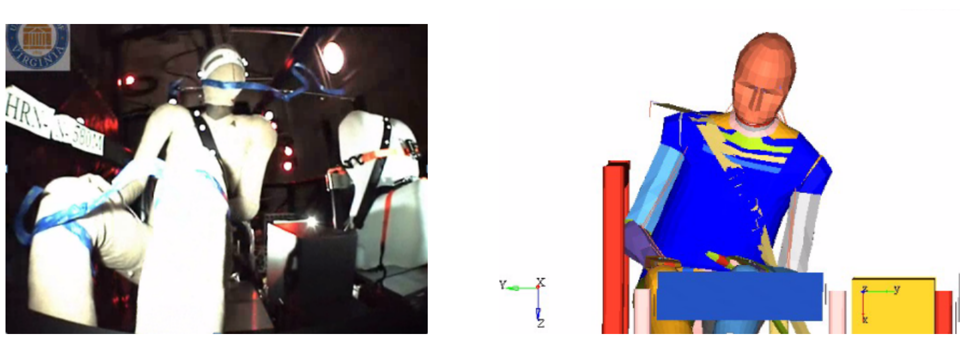
Figure 14: Front Comparison t=0.5 s

Figure 15: Front Comparison t=1s

In the X- direction (Occupant forward in seat is positive), the Active Human Model showed more discrepancies with respect to the corridors. As seen in Figure 16.

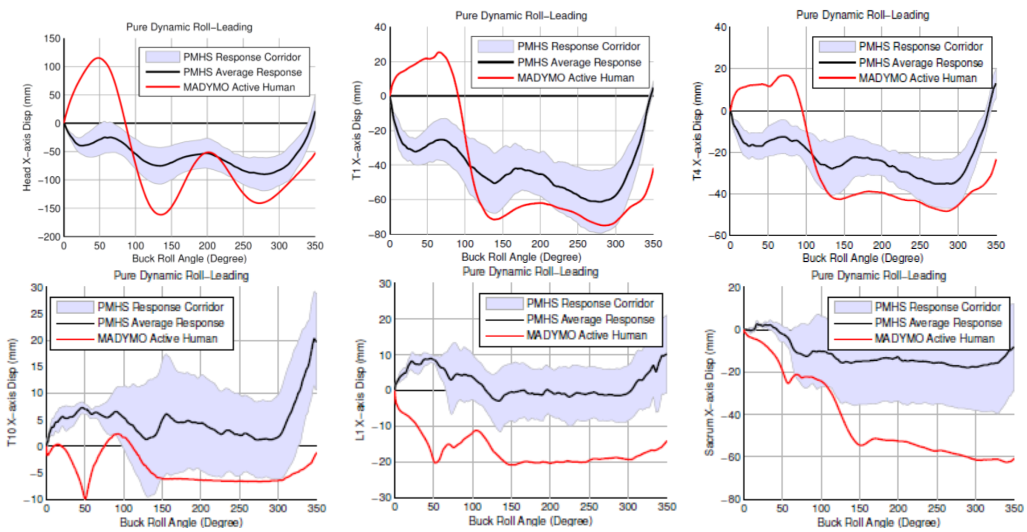

Figure 16: Corridors in the X Direction

The model not responding as well in the X-direction is to be expected since that direction is the non-loading path in this rollover scenario. The model will be more sensitive to any small difference between the Active Human Model and the PMHS.

The Effect of Adding the Tape

The effect of adding tape to the head can be seen in Figure 17. Adding the tape did not change the initial direction that the head moved. The head still moved forward as opposed to the PMHS, which moved backwards initially. The tape did however decrease the distance between the PMHS and the Active Human Model in the X direction for the head.

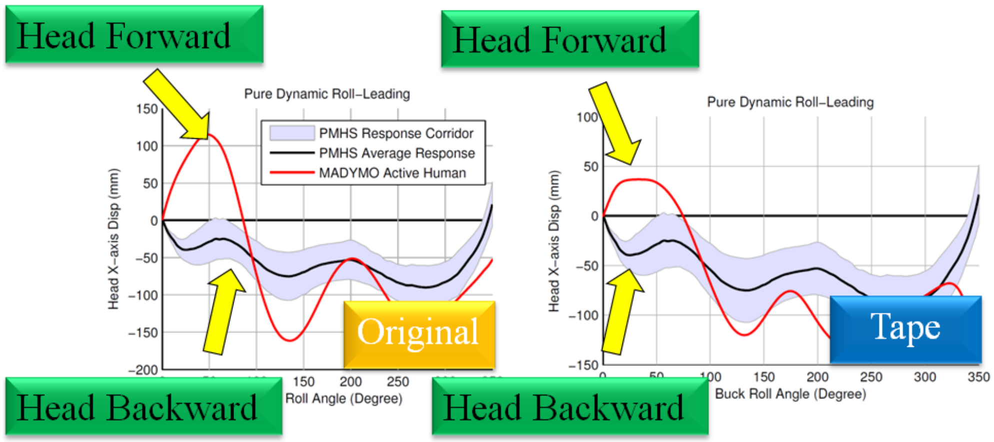

Figure 17: Head X-Direction Corridors With and Without Tape

Investigation into the Pelvis Sliding

As seen in the corridors in Figure 12, the pelvis of the Active Human Model goes a farther distance outboard then that of the PMHS. The pelvis kinematics, from comparing the video of the PMHS to that of the simulations run, greatly differs. The pelvis of the Active Human Model is the first of the frontal plane to move outboard. In the PMHS, the pelvis is the last to go outboard.

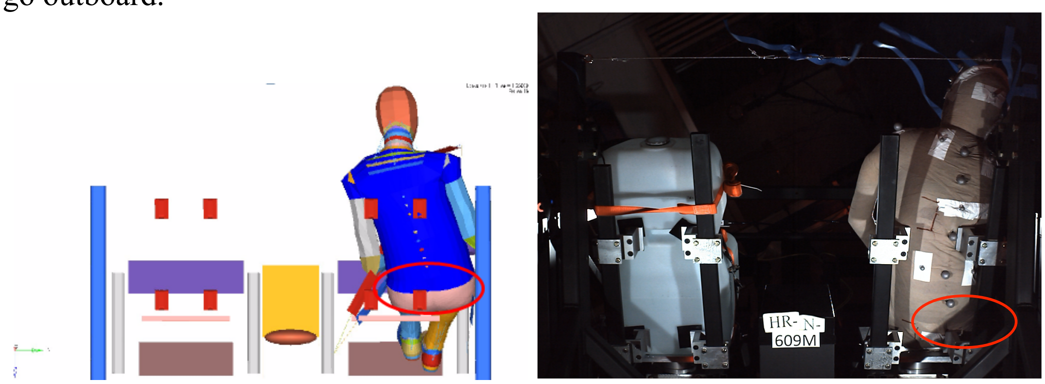

Figure 18: Difference in Pelvis Kinematics

This pelvis issue was thought to be a boundary condition problem, so incredibly high friction was tried initially to see if it would hold the pelvis from going outboard. This did not seem to work as intended as the pelvis lifts off of the seat and so is not constrained from going outboard. The vertebrae bending stiffness characteristic in the lateral direction were then scaled up so to see if that was a possible cause. Scaling up the stiffness of the thoracic spine did seem to have an effect in making the Active Human Model look closer to the PMHS response but since those characteristics are encrypted, the remaining simulations were run with the original non-scaled characteristics.

Sensitivity to Different Boundary Conditions

Changing of the Initial Position

Corridors were created for the baseline position. As seen in Figure 19, different positioning does not have a great effect in the y- direction for the upper body.

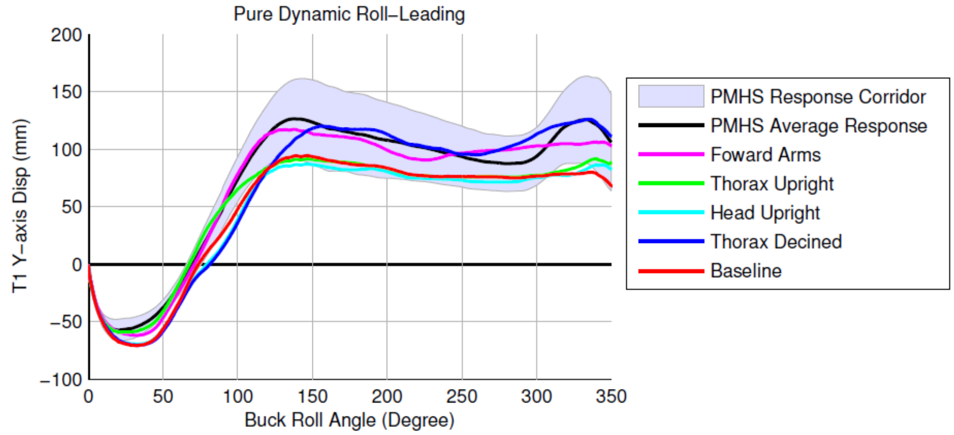

Figure 19: T1 Y-axis Corridors of Differently Positioned Models

However, the farther down the vertebrae, the more some of the different positions had an impact. Moving the arms forward and declining the thorax had a consequence on the lower region. This is because these two scenarios created a situation where there was a torque about the models Z-axis which changed the interaction between the models right shoulder and that of the buck, allowing for the model to slide more outboard.

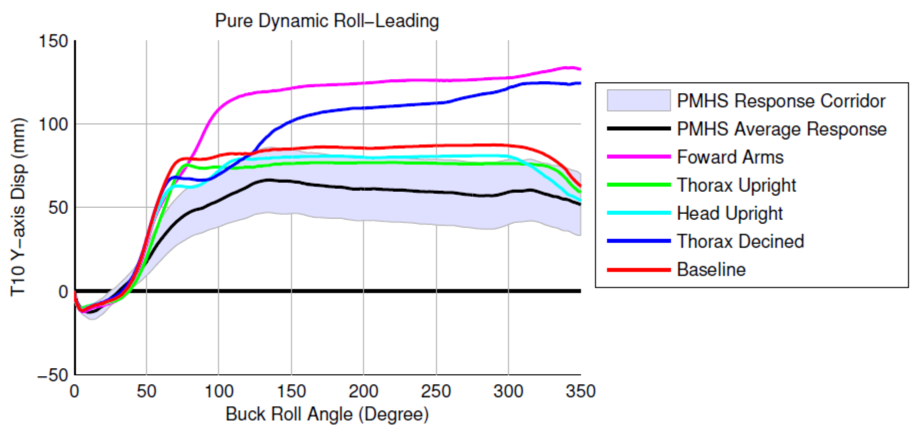

Figure 20: T10 Y-axis Corridors of Differently Positioned Models

The declined thorax also allowed for the lower body to skid out from under the lap belt and create bigger excursion in the X and Z directions.

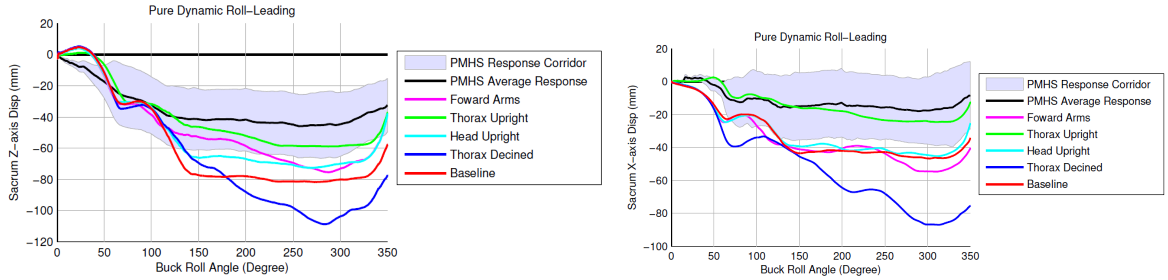

Figure 21: X and Z Displacement of the Sacrum at Different Positions

Changing the Seat Friction Coefficients

The overall pattern of increasing the seat friction is that the Active Human Model is better coupled to the seat. So there are fewer excursions when the seat friction is higher. This holds less true the higher up on the body the marker is. How the head is loaded creates a situation where the friction does not have much of an effect on the movement of the head. This is especially true in the Y-axis direction.

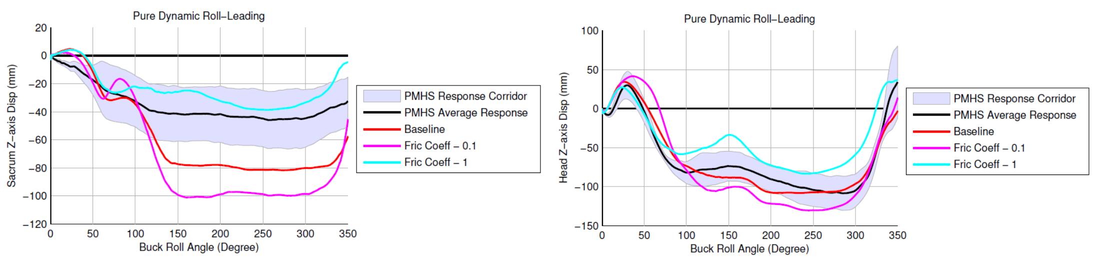

Figure 22: Friction Coefficient Effects on Upper and Lower Z-Axis Displacement

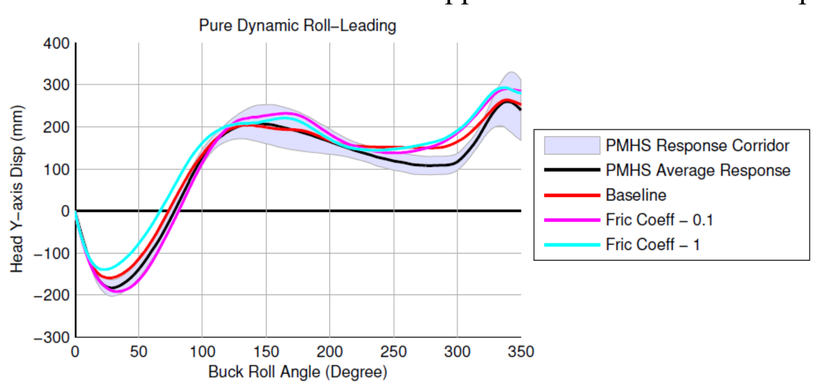

Figure 23: Friction Coefficient Effects on the Y-Axis Head Displacement

Changing the Seat Belt Tension

Increasing the belt tension had a similar effect to that of increasing the seat friction. The higher the belt tension, generally the greater the occupant is coupled to the motion of the buck. The major difference between changing the seat friction and the belt tension is that a lower belt tension led to more excursion of the Active Human Model in the X-axis direction as shown in Figure 24.

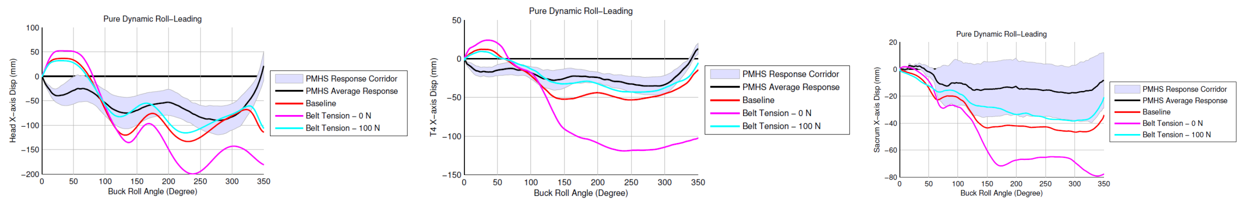

Figure 24: X-Axis Displacement of the Head, T4, and Sacrum at different Belt Tension Loads

The head's movement was little affected by the belt tension that was prescribed. The head was more a function of the loading from the revolution of the buck. This is seen clearly in Figure 25.

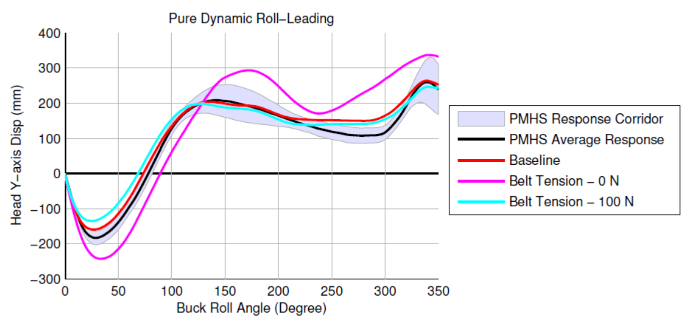

Figure 25: Head Y-Axis Displacement for Different Belt Tensions

The result of using different belt tensions is that minimal belt tension has more of an effect on the occupant kinematics then greater belt tension.

DISCUSSION

The occupant kinematics are important to understand where the orientation and location of the occupant relative to the vehicle. This is significant in understanding how the impact interacts with the occupant. Most of the injuries that occur in rollover crashes occur as compressive loads to the head and cervical spine. So it can be argued that the kinematics of the head are the most important when evaluating the model.

The upper body of the Active Human Model was closer to the baseline response of the PMHS then that of the six ATDs presented by Zhang et al. (2014). This is especially apparent in the loading directions of the buck on the Y and Z axis. Figure 26 shows the comparison of the Active Human Model to that of the ATD for the T1 marker along the Y-axis.

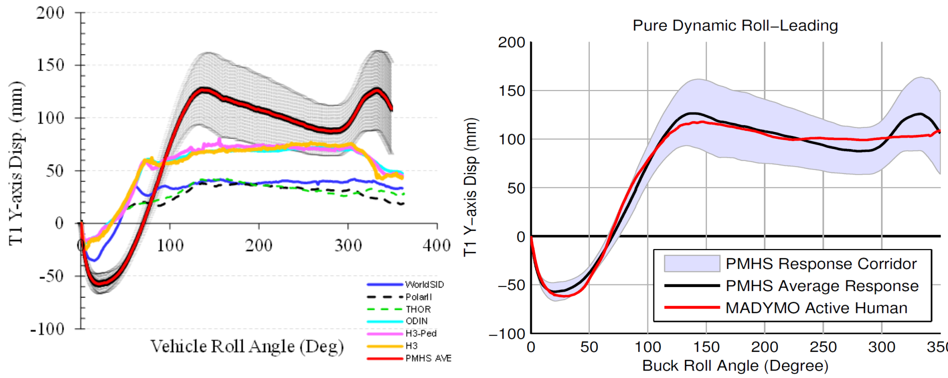

Figure 26: Kinematic Corridors of ATD and Active Human Model for T1 Y-Axis

It is worth noting that even though the initial direction of the Active Human Model head along the x-axis is opposite of the PMHS head, the head does later go on in the roll in the correct direction. At the range of possible touchdown conditions, the head of the Active Human Model is still relatively close to where the PMHS head is located in the X-axis direction. When looking at corridors that were published by Zhang et al. (2014) for the dummies subjected to a similar condition as the PMHS tests, the head of the dummies go the opposite of the PMHS initially and never return to the same direction as the PMHS. Figure 27 shows where the head will be for the PMHS, Active Human Model and dummies will be in the range of 150 degrees, which is in the range of possible touchdown conditions.

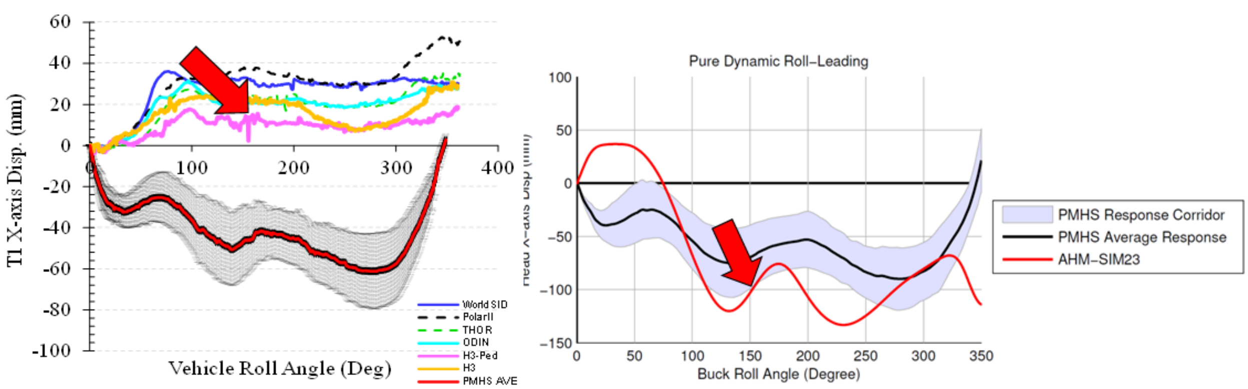

Figure 27: Differences in PMHS, ATDs, and the Active Human Model at Impact Range

The result of changing the positioning revealed that the interaction of the shoulder with the outside of the buck determined how the Active Human Model would rotate about its Z-axis. This occurred while moving the model forward so that that thorax was at more of a decline then the model and when the arms of the model were moved forward. The other positions did not have as much of an effect on the kinematic response. While seat friction and seat belt tension had similar overall effects when coupling the occupant to the buck, decreasing the initial belt tension seemed to have a greater effect. The less tense belt allows for more excursions not only in the Y and Z axis directions but the X direction which changes how the occupant interacts with the back of the seat all together.

The response of the lower portion of the model did not improve when changing the different boundary conditions. This may be due to how the PMHS interacted with the back block that could not be properly modeled or to the material characteristics for the lateral bending in the spine. Further investigation must be taken in order to fully understand what is occurring. Future work also includes turning on the active component of the model and comparing possibly to some volunteer data.

CONCLUSIONS

The goal of this study was to evaluate the kinematic response of the MADYMO Active Human Model to that of PMHS under controlled rollover conditions. The results of which show that the MADYMO Active Human Model upper body response in the YZ lateral plane is closer to PMHS in a rollover event then that of six of the leading crash test dummies. The imposed (vehicle) kinematics and restraint resulted in a low sensitivity of initial position to peak lateral and vertical excursion except in cases where the shoulder interaction with the buck was changed. Likewise, lowering the belt tension resulted in relatively large changes in kinematic response. Though there is some trepidation with the response of the lower body the overall model has been shown to have satisfactory kinematic biofidelity in the atonic configuration, it will be used to examine the effect of variations in active muscle tensing to lend insight into protecting humans in rollover crashes.

ACKNOWLEDGEMENTS

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