

Finite-Element-Based Transfer Equations: Post-Mortem Human Subjects versus Hybrid III Test Dummy in Frontal Sled Impact

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ABSTRACT

Transfer or response equations are important as they provide relationships between the responses of different surrogates under matched, or nearly identical loading conditions. In the present study, transfer equations for different body regions were developed via mathematical modeling. Specifically, validated finite element models of the age-dependent Ford human body models (FHBM) and the mid-sized male Hybrid III (HIII50) were used to generate a set of matched cases (i.e., 192 frontal sled impact cases involving different restraints, impact speeds, severities, and FHBM age). For each impact, two restraint systems were evaluated: a standard three-point belt with and without a single-stage inflator airbag. Regression analyses were subsequently performed on the resulting FHBM- and HIII50-based responses. This approach was used to develop transfer equations for seven body regions: the head, neck, chest, pelvis, femur, tibia, and foot. All of the resulting regression equations, correlation coefficients, and response ratios (FHBM relative to HIII50) were consistent with a set of test-based results. The HIC15 transfer equation was used to transform the cadaver-based risk curve into the HIII50 domain.

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INTRODUCTION

There were 33,561 road user fatalities in the United States of America in 2012. Among them, 12,271 (37%) were passenger car occupants. Between 1970 and 2012, the US experienced 60% reduction in passenger car occupant fatalities. Further reductions are expected with additional advancements in restraint technologies and electronic stability control, increase in the use of safety belts and age-appropriate child safety seats, and continued decrease in alcohol- and drug-impaired driving (NHTSA, 2014). In 2012, fatalities in frontal crashes constituted 61.8% of the total number of fatalities in all crashes and injuries caused by frontal crashes made up 70.9% of injuries in all crashes (NHTSA, 2012).

Statistical analyses are typically used to derive risk curves. Risk curves relate cadaveric injury outcomes such as bone fracture to corresponding physical input variables such as force, compression, etc. Injury outcomes may be obtained from tests involving postmortem human subjects (PMHS). The responses of test dummies like the Hybrid III mid-sized male dummy (HIII50) are often directly compared to cadaver-based risk curves. Response or transfer equations that relate the cadaveric responses to those of the HIII50, subjected to matched or nearly-identical loading conditions, may potentially improve the fidelity and applicability of those risk curves.

One method of relating PMHS- to dummy-responses is through testing both surrogates under matched loading conditions. Test-based transfer equations are often difficult to generate due to the limited availability of PMHS for matched testing. Laituri et al. (2005) derived a test-based transfer equation. In that study, the test-based transfer equation was used to convert the PMHS-based risk curve into a HIII50-deflection-based risk curve. However, a literature search resulted in 20 approximately-matched tests. Many of those tests involved mixed loading conditions. In addition, due to the limited number of those tests, the effect of age could not be established from the matched tests. Laituri et al. subsequently applied the resulted, albeit limited, test-based transfer equation. The authors of that study noted that supplemental math modeling might be useful.

In the present study, model-based transfer equations were developed. First, finite element (FE) models of the age-dependent Ford Human Body Model (FHBM) and the HIII50 were validated against physical tests. Next, the validated models were used to create a set of responses from matched cases involving different impact speeds, severities, and FHBM ages. The resulting FHBM and HIII50 responses were then subjected to regression analyses to generate transfer equations. This approach was applied in separate analyses of the head, neck, chest, pelvis, femur, tibia, and foot. Finally, those model-based transfer equations were compared to test-based transfer equations that were developed, using the same regression analyses, from limited published matched-condition tests of the same surrogates tested under nominally identical loading conditions.

METHODS

The age-dependent FHBMs and Altair's HIII50 Radioss models were used in the current study. The models were previously validated against published tests. Those tests included frontal blunt and sled impacts. Details on the tests and validation results can be found in (El-Jawahri et al. 2014).

The FHBM and the HIII50 FE models were subjected to matched-condition simulations of frontal sled impacts at different severities and speeds. The sled had dimensions based on the average of a number of mid-sized sedans (Backaitis et al. 1995). It had a rigid seat, a padded knee-bolster, a deformable steering wheel, and a rigid foot rest. The knee-bolster padding thickness and material parameters represented a typical production vehicle. The steering column stroke was modeled, but only translation was allowed. [Figure 1](#) shows the content and dimensions of the sled. Each surrogate was restrained with a 3-point standard belt with and without a single-stage-inflator airbag. Belt elongation was 12%. The belt anchorage points were the same for all surrogates. The shoulder belt was routed so that it passed over the mid-clavicle and mid-sternum of each surrogate. [Figure 2](#) shows the belt orientation on each surrogate.

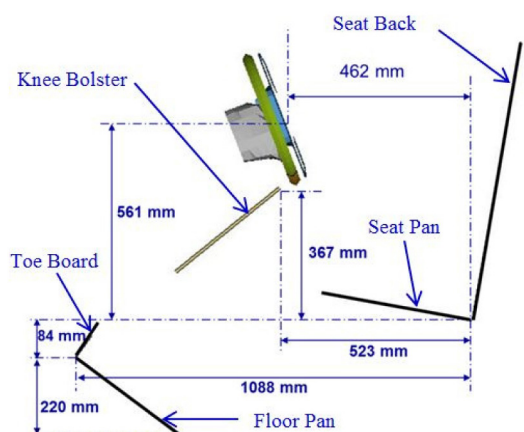


Figure 1. Sled content and geometry

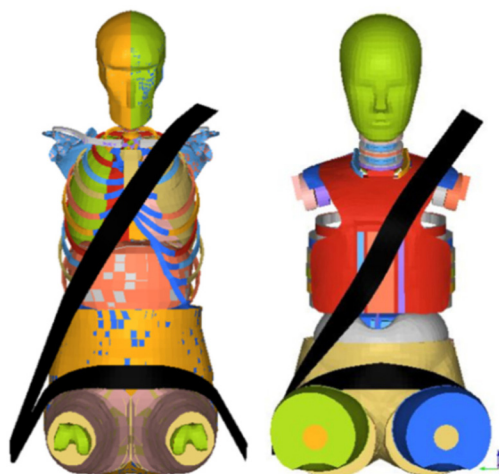


Figure 2. Belt orientation: left FHBM, right HIII50

In a previous study (El-Jawahri et al. 2014), the FHBMs and HIII50 models were subjected to matched blunt impact conditions. The two surrogates were positioned such that the orientations of the body region of interest were similar. In the current study, each surrogate was positioned to fit the sled. As a result, the orientations of some body regions were different. For example, the head and chest of the HIII50 were closer to the steering wheel compared to those of the FHBMs. The pelvises, femurs, legs, and feet of the two surrogates were oriented similarly. Overlays of the two positioned surrogates are shown in [Figure 3](#) (Blue for FHBM and Red for HIII50).

CHEST DEFLECTION

The external chest deflection of the age-dependent human models and the slider deflection of the HIII50 were gathered from the respective simulations. The external chest deflections of the FHBMs were chosen because in most of the cadaveric tests those are typically measured. Those chest deflections are typically normalized by the depth of the respective cadaver chest. In the current study, the chest deflections of the FHBMs and HIII50 were normalized by the depth of the respective chest of each surrogate. The FHBM chest depth was measured from the model at the 4th intercostal space in the mid-sagittal plane (213 mm). The HIII50 chest depth was 229 mm (Neathery et al. 1974).

SLED MODEL VALIDATION

The sled model was validated against three different test datasets. The first dataset consists of unbelted HIII50 frontal sled tests that were run with ΔV s of 10, 12, 14, 16, and 18 mph (Laituri et al. 2006). The second set represents the average of multiple 30 mph CMVSS 208 tests ($n = 48$) where the HIII50 was restrained with a standard belt only. The third set is based on 30 mph CMVSS 208 tests ($n=64$) where the HIII50 was restrained with a standard belt and an airbag. For each case, the responses of the HIII50 model were compared to the respective test data. More details on the CMVSS 208 data can be found in Laituri et al. 2003a and 2003b.

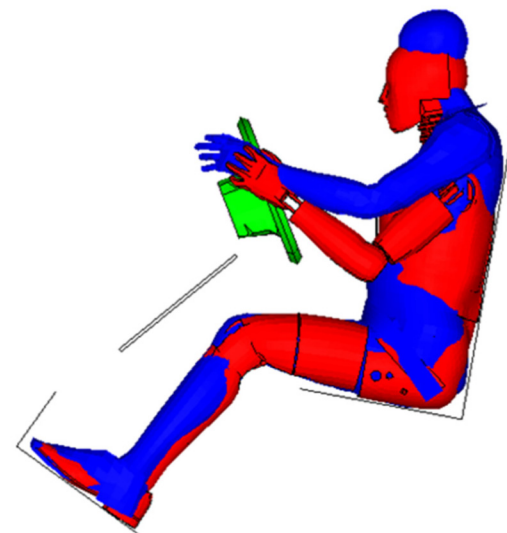


Figure 3. Surrogate orientation in sled: blue FHBM, red HIII50

SLED PULSES

Pulses with different severities and speeds were used. Specifically, 26 different pulses were used with each occupant. The pulses were classified into three severity groups: non-potentially barrier like event (NPBLE), potentially barrier like events (PBLE), and new car assessment program pulses (NCAP). There were 7 NPBLE and 7 PBLE pulses both increasing from 5 to 35 mph ΔV in 5 mph increments and 12 NCAP pulses at 35 mph ΔV . More details on the PBLE and NPBLE pulses can be found in Laituri et al. 2003a. The matched-condition simulations were run for each surrogate with all the pulses for the belt-only cases. Since driver airbags are typically fired at around 15 mph ΔV or higher, the 22 pulses with ΔV greater than 15 mph were simulated for the belt-and-airbag cases.

AIRBAG FIRING TIME

The time for firing the airbag was determined subject to a “5 inches minus 30 milliseconds” methodology. The firing time is calculated by subtracting 30 ms from the time needed for an unbelted HIII50 occupant to travel a distance of 5 inches. The firing time calculation is a pulse-dependent phenomenon. MADYMO models for each of the 26 pulses were run to determine the airbag firing times based on the aforementioned method.

MODEL-BASED TRANSFER EQUATIONS

The corresponding model responses of all surrogates were obtained from the simulation output. The responses were: peak resultant acceleration of head center of gravity (cg), HIC15, peak upper neck shear, peak upper neck tension-compression force, peak upper neck flexion-extension moment, peak N_{ij} , peak chest acceleration, chest acceleration 3-ms clip, peak chest deflection, peak pelvis acceleration, peak femur axial force, peak upper tibia axial force, peak upper tibia x- and y-moments, peak lower tibia axial force, peak lower tibia x- and y-moments, peak foot acceleration, and peak shoulder belt force. Regression analyses were performed on the resulting FHBM- and HIII50-model-based responses. The resulting regression equations were called “model-based transfer equations.”

TEST-BASED TRANSFER EQUATIONS

There are some published studies with the two surrogates tested under nominally-identical conditions. To compare the model-based responses, the PMHS and HIII50 responses from those tests were gathered. The loading conditions of those tests differed. Due to the limited number of exactly matched-condition tests, the test-based responses were not grouped subject to the type of loading or restraint system used. Those studies included the following:

1. Frontal sled tests with the HIII50, THOR, and cadavers (Petitjean et al. 2002). In that study, two types of restraint systems were used: a 6 kN load-limiting 3-point belt, and a 4 kN load-limiting belt with an airbag. A Euro NCAP pulse that represents a 64 km/h impact against a deformable barrier with 40% overlap was used. A knee bolster and steering wheel were present in those tests.

2. Frontal sled tests with the HIII50, THOR, and cadavers at 30 and 50 km/h (Vezin et al. 2002). The surrogates were seated in a rigid seat similar in geometry to that of a mid-sized car. A three-point belt with a 4 kN shoulder belt load limiter was used in the 30 km/h tests. The restraint used in the 50 km/h tests consisted of a 3-point belt with a 4 kN shoulder belt load limiter and an airbag.
3. Several studies of matched-condition tests with PMHS and HIII50 at different loading conditions were cited by Laituri et al. (2003a and 2003b). In those two studies chest deflections of both surrogates were reported. More details on the different studies can be found in Laituri et al. (2003a and 2003b).
4. Frontal sled tests with the HIII50, THOR, and cadavers at 29 and 38 km/h (Forman et al. 2006). The surrogates were seated in a rigid seat of a mid-sized sedan sled buck and restrained with a standard 3-point belt.

The relevant responses from the two surrogates were collected. Subsequently, regression analyses were performed to derive test-based transfer equations. Similar to the model-based approach, a linear fit that passes through the origin was assumed.

RESULTS

For brevity, the 55 YO FHBM relative to the HIII50 response plots for the head and chest responses will be presented. The transfer equations for the other responses will be summarized later. The results will be grouped into three parts: the sled model validation, derivation of the head response transfer equations, derivation of the chest deflection transfer equation, and transfer equations for the rest of the responses.

SLED MODEL VALIDATION

Figure 4 compares the test and model results from the frontal sled impact with the unbelted HIII50 at different speeds. The model responses and the respective fleet averages of CMVSS 208 are shown in Figures 5 and 6. For easier presentation, each response was normalized to its respective injury criterion performance limit (ICPL) as defined by FMVSS 208. The ICPL values for HIC36, chest acceleration, chest deflection, and femur load are 1000, 60 g, 63 mm, and 10 kN, respectively. The ICPL for the shoulder belt was determined from the chest deflection ICPL. It is based on the following: each 0.2 kN of shoulder belt load results in 1 mm of chest deflection (Mertz et al. 1991). Since the ICPL for chest deflection was 63 mm; the calculated shoulder belt load ICPL was 12.6 kN. The figures show that the model was in good agreement with the test and fleet data for the different restraints and speeds. The chest deflection for the unrestrained occupant was under-predicted by the model.

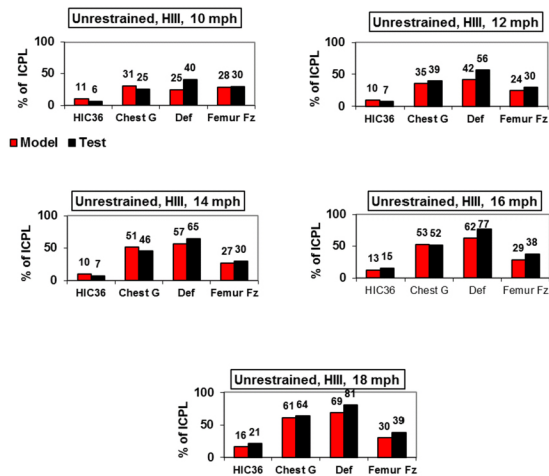


Figure 4. Comparison of test/model sled results (unrestrained)

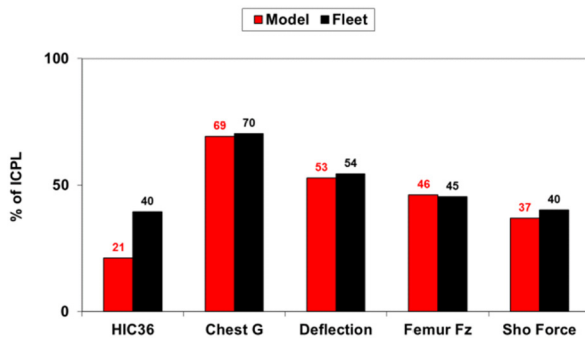


Figure 5. HIII50 sled model validation, 30 mph ΔV (belt-and-airbag)

HEAD RESPONSE TRANSFER EQUATIONS

The peak head cg resultant acceleration and HIC15 were gathered from the matched-condition simulations of the two surrogates.

Regression analyses were performed for three datasets subject to the restraint type: all data, belt-only cases, and belt-and-airbag cases. This resulted in three transfer equations, one for each dataset. Figures 7 and 8 show the peak head acceleration of the 55 YO FHBM and HIII50 for all cases (n = 48) and belt-and-airbag (n = 22), respectively. There were only 22 belt-and-airbag cases because the airbag did not deploy for speeds below 15 mph. Consequently, the 5 and 10 mph belt-and-airbag cases were not considered. Linear relationships that pass through the origin were assumed between the responses while deriving the transfer equations for the different body regions. The figures also show the respective test-based matched data of the two surrogates. Due to the paucity of the test data, those data were not classified subject to the restraint type. Figures 9 and 10 show the HIC 15 values of both surrogates for all cases and belt-and-airbag simulations, respectively.

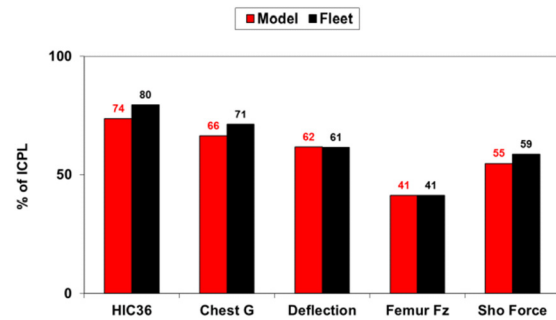


Figure 6. HIII50 sled model validation, 30 mph ΔV (belt only)

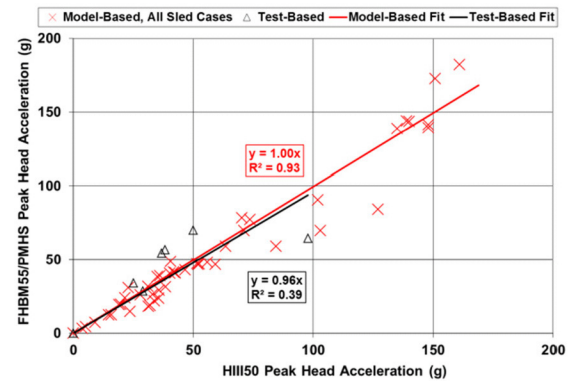


Figure 7. 55 YO FHBM and HIII50 head acceleration (all cases)

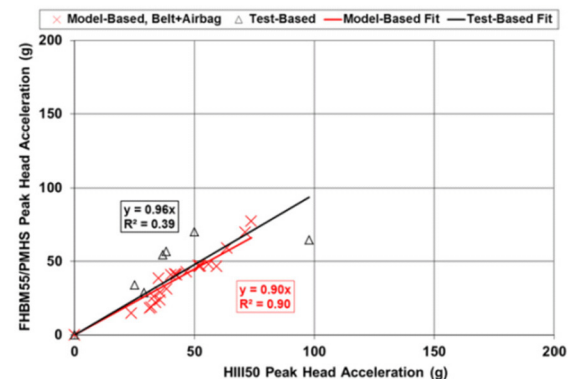


Figure 8. 55 YO FHBM and HIII50 head acceleration (belt+airbag)

CHEST RESPONSE TRANSFER EQUATIONS

The external chest deflection at the 4th intercostal space of the age-dependent FHBMs and the chest deflection at the slider of the HIII50 were gathered from the matched-condition simulations of the two surrogates. The chest deflections of the FHBMs and HIII50 were normalized to the respective chest depths at the location of the chest deflection measurement. The chest depth of the FHBMs is 213 mm. The HIII50 chest depth is 229 mm (Neathery et al. 1974). Figures 11 and 12 show the normalized chest deflections of the 55 YO FHBM and HIII50 for all cases (n = 48) and belt-and-airbag (n = 22), respectively. The figures also show the test-based matched data of the two surrogates. Due to the paucity of the test data, those data were not classified subject to the restraint type.

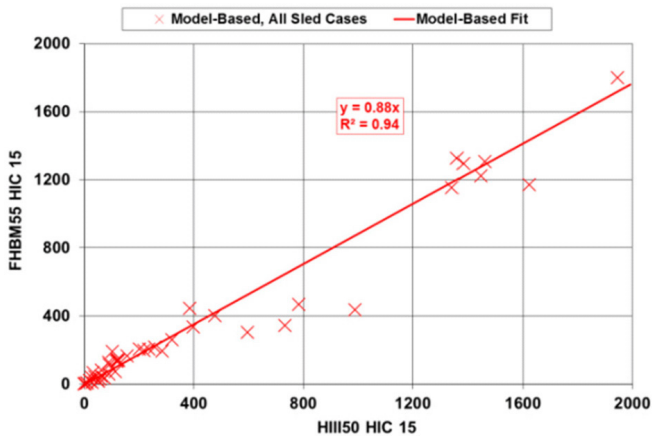


Figure 9. 55 YO FHB55 and HIII50 HIC 15 (all cases)

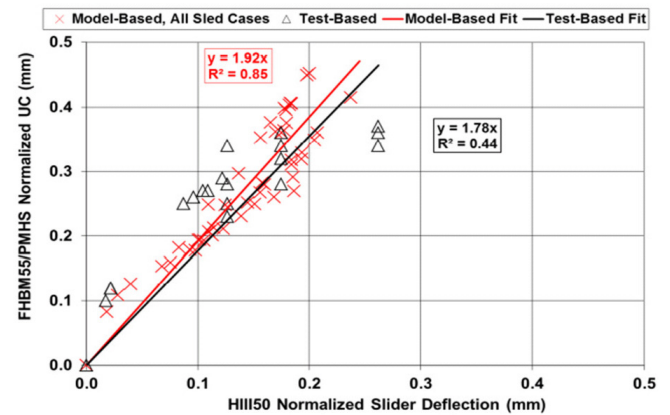


Figure 11. 55 YO FHB55 and HIII50 normalized deflection (all cases)

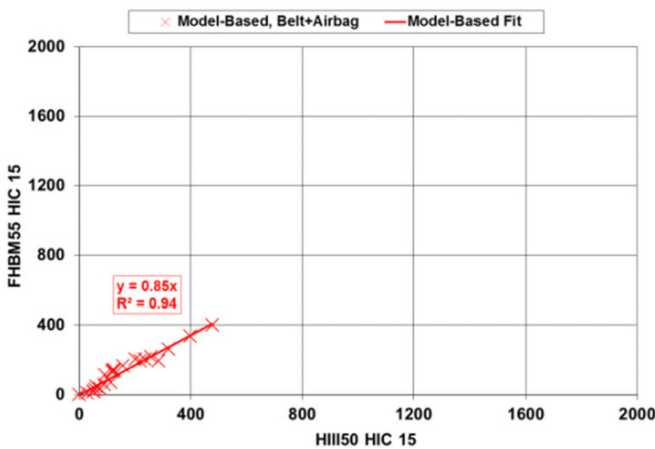


Figure 10. 55 YO FHB55 and HIII50 HIC 15 (belt-and-airbag)

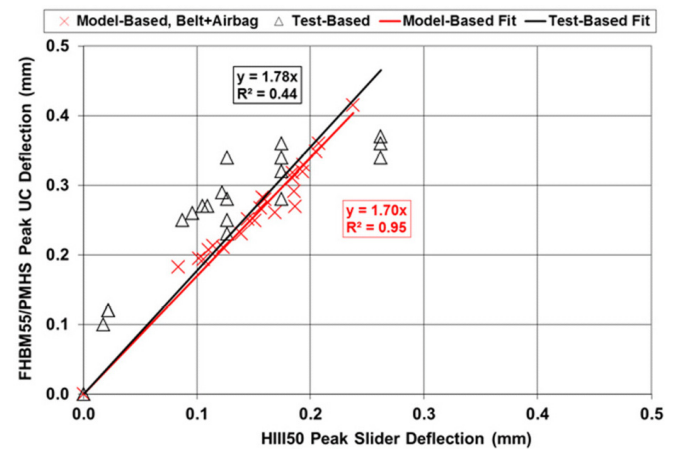


Figure 12. 55 YO FHB55 and HIII50 normalized deflection (belt-and-airbag)

MODEL-BASED TRANSFER EQUATIONS

Similar to the head regression analyses, linear relationships that pass through the origin were assumed between the age-dependent FHBMs and HIII50 responses for all body regions. Figures A1, Figures A2, Figures A3, Figures A4, Figures A5, Figures A6, Figures A7, Figures A8, Figures A9, Figures A10, Figures A11, Figures A12, Figures A13, Figures A14, Figures A15, Figures A16 illustrate the model-based peak responses of the FHBMs and HIII50. The figures also show the respective test-based data (if available). Tables A1, A2, A3, A4 list the transfer equation slopes for the model-based FHBMs and HIII50 responses. The respective R^2 values are also listed in those tables. For each of the upper body responses, three transfer equations were derived subject to the loading type. Those are: all sled cases, belt-and-airbag cases, and belt-only cases. For lower body responses, it was assumed that an airbag would have little to no effect on those responses.

MODEL- AND TEST-BASED TRANSFER EQUATIONS

As mentioned earlier, test-based responses of PMHS and HIII50 under nominally-identical conditions were gathered from literature. The same regression analyses used in the model-based data were performed on the test-based data. Table A5 lists the dataset size, slope of the linear fit, and R^2 values of the test-based transfer equations for the different responses. The table also lists the respective dataset size, slope and R^2 values of the model-based transfer equations.

DISCUSSION

Head Response Transfer Equations

As seen from Figures 7 and 8, the model-based peak head acceleration of the 55 YO FHB55 was similar to that of the HIII50 when all cases were considered. However, in the belt-and-airbag cases the peak head acceleration of the 55 YO FHB55 was lower than that of the HIII50 by 10%. The figures also show that the test-based data were within the spread of the model-based data. The test-based peak head acceleration of the PMHS was lower than that of the HIII50 by 4%.

The model-based HIC 15 of the 55 YO FHBM was 12% lower than that of the HIII50 when all cases were considered. The HIC 15 of the 55 YO FHBM was 15% lower than that of the HIII50 when only the belt-and-airbag cases were considered. No matched-condition test data are available. The aforementioned differences between the head responses of the two surrogates may be attributed to the deformability of the human head compared to the rigid dummy head and/or the differences between the neck and spine of the two surrogates. Figures 13 and 14 show the head acceleration time histories of both surrogates in the 25 and 35 mph ΔV PBLE for the belt-and-airbag and belt-only cases, respectively. In the belt-and-airbag cases the peak head accelerations of both surrogates were comparable for both speeds however; the timing and duration of the peaks were different. The same was observed in the lower speed belt-only case (i. e. comparable peaks but different timing and duration). In the higher speed belt-only case, the head of the FHBM and HIII50 contacted the steering wheel at 95 and 86 ms, respectively. The location of contact on the steering wheel was also different. Figures 15 and 16 show the head location of both surrogates relative to the airbag and steering wheel at the time of contact for the belt-and-airbag and belt-only cases for the 35 mph ΔV PBLE cases, respectively.

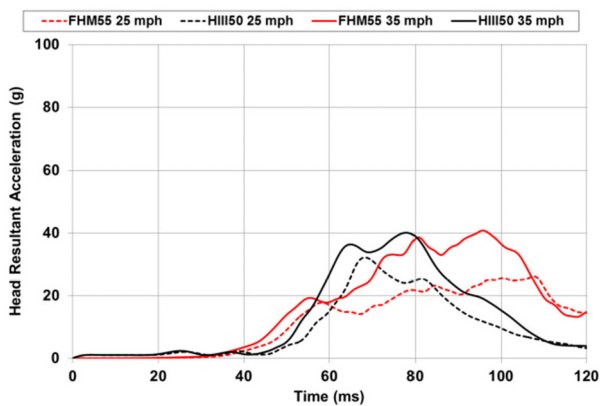


Figure 13. 55 YO FHBM and HIII50 head acceleration (belt-and-airbag)

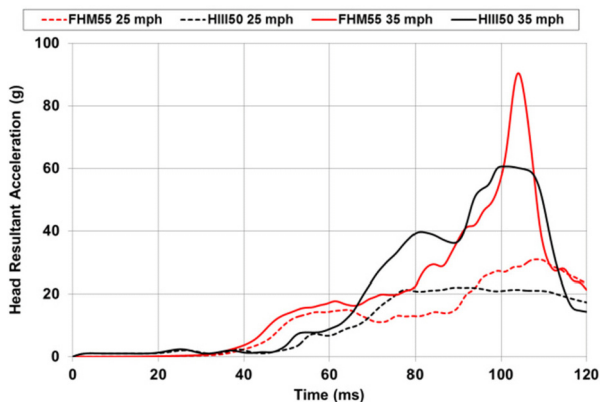


Figure 14. 55 YO FHBM and HIII50 head acceleration (belt only)

Figures 17 and 18 show the y-x and z-x trajectories of the head for the belt-and-airbag and belt-only of the 35 mph ΔV PBLE cases, respectively. The x-displacements of both heads were comparable for both loading cases. However, the y- and z-displacements of the

FHBM head were greater compared to those of the HIII50 in both loading cases. The FHBM head moved about 150 and 100 mm to the left side of the occupant in the belt-and-airbag and belt-only cases, respectively. Whereas, the head of the HIII50 moved about 5 and 10 mm to the right side of the occupant in those cases, respectively. With respect to the Z-axis, the FHBM head moved 105 and 200 mm downward in the belt-and-airbag and belt-only cases, respectively. Whereas, the HIII50 head moved about 40 and 150 mm downward in those cases, respectively.

Neck Responses Transfer Equations

The 55 YO FHBM neck responses were lower than those of the HIII50. The neck shear, tension, and moment were 62, 53, and 91% lower than those of the HIII50 when all loading cases were considered, respectively. Whereas, those responses were 50, 51, and 90% lower than those of the HIII50 when only belt-and-airbag cases were considered, respectively. The 55 YO Nijs were 80 and 68% lower than those of the HIII50 when all loading cases and belt-and-airbag cases were considered, respectively (see Figures A1, A2, A3, A4). This could be explained by the differences in necks of the two surrogates. The human neck has deformable vertebrae and vertebral discs. The vertebrae are connected by spinal ligaments. The head is attached to the neck through ligaments and muscles. The HIII50 neck is made of aluminum parts and rubber disks to represent the vertebral bodies and the intervertebral discs, respectively. The muscle of the neck is represented by a central steel cable. The head is rigidly connected to the HIII50 neck.

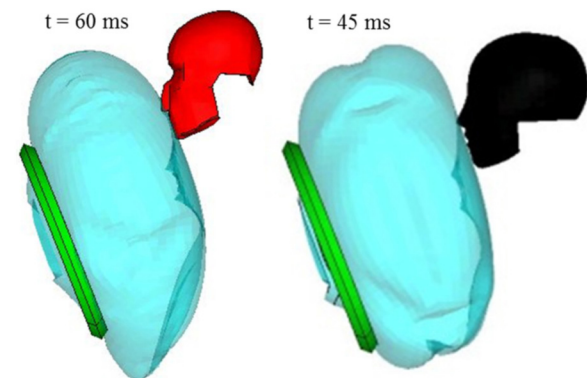


Figure 15. 55 YO FM and HIII50 head at time of contact 35 mph ΔV PBLE: belt/airbag (left), belt only (right)

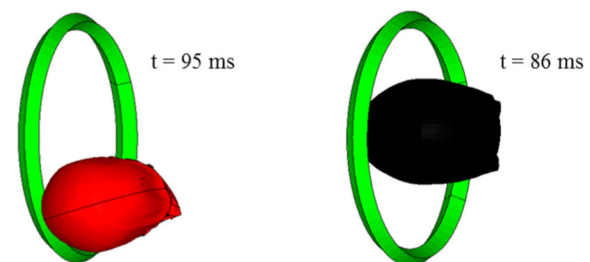


Figure 16. 55 YO FM and HIII50 head at time of contact 35 mph ΔV PBLE (belt only)

Chest Transfer Equations

As Figures 11 and 12 illustrate the 55 YO FHBM normalized chest deflections were 92% and 70% higher than those of the HIII50 when all loading cases and belt-and-airbag cases were considered, respectively. When compared to the test-based data, the model responses were within the spread of the test data. However, the belt-and-airbag cases were closer to the test data. The human ribcage has 12 ribs slanted at different angles with respect to the thoracic spine.

The first rib and clavicle are connected to the top of sternum. The HIII50 ribcage has 6 elliptical horizontal ribs. There is no clavicle-to-sternum connection in the HIII50 ribcage. The HIII50 is instrumented with a single-point chest deflection measurement at the mid-sternum level. This location was chosen to represent the 4th intercostal space in humans.

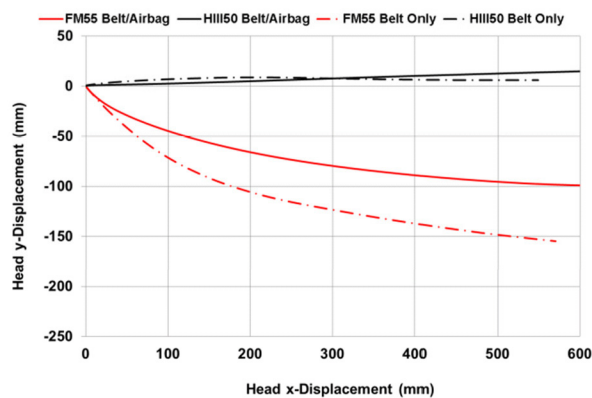


Figure 17. 55 YO FHBM and HIII50 head trajectories 35 mph ΔV PBLE (y-x)

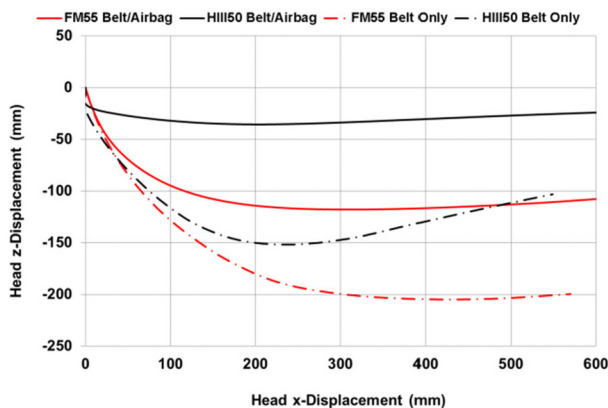


Figure 18. 55 YO FHBM and HIII50 head trajectories 35 mph ΔV PBLE (z-x)

Two additional transfer equations were developed, namely: the 3 ms acceleration clip and the peak chest resultant acceleration. The 55 YO FHBM 3 ms acceleration clips and peak chest accelerations were 21 and 44% higher than those of the HIII50, respectively when all loading cases were considered. The same responses were 21 and 41% higher than those of the HIII50, respectively, when the belt-and-airbag cases were considered. The test-based PMHS 3 ms accelerations were 5% higher than those of the HIII50. There was no

published test data reporting the peak chest accelerations of the two surrogates under the same loading conditions (see Figures A5 and A6).

Pelvis and Femur Transfer Equations

For the lower body responses, it was assumed that the airbag has little effect on the lower body responses. As such, only all loading cases were considered in deriving the transfer equations for those regions. Three transfer equations were derived: pelvis peak resultant acceleration, left femur F_z , and right femur F_z . The 55 YO FHBM peak pelvis accelerations were similar to those of the HIII50 when all loading cases were considered. When compared to the test-based data, the PMHS peak pelvis accelerations were 1% lower than those of the HIII50 (see Figure A7).

It was observed that the left femur F_z of both surrogates were higher than those of the right femur. The hypothesis is that the driver belt routing tends to restrain the right side more than the left side of the occupant since the shoulder belt goes on the right side of the occupant whereas, the upper left side of the occupant is relatively unrestrained. The 55 YO FHBM peak left femur axial forces were 85% lower than those of the HIII50. Whereas, the 55 YO FHBM right femur axial forces were 81% lower than those of the HIII50. There was no published test-based data reported femur axial forces of the two surrogates under the same loading conditions (see Figure A8). In addition to the pelvis and femurs being deformable in humans and rigid in the HIII50, there are some geometrical differences in the femur that can result in differences between the pelvis and femur responses. Figures 19 and 20 illustrate the geometrical differences between the human and HIII50 lower extremities. There are three differences in the femur of the human relative to that of the HIII50: 1) it is longer, 2) it has bigger eccentricity between the femur head and the mid-shaft, and 3) the curvature is more pronounced.

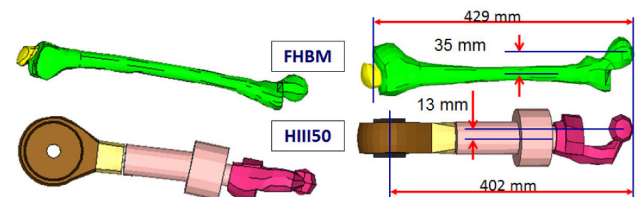


Figure 19. FHBM and HIII50 femur comparison

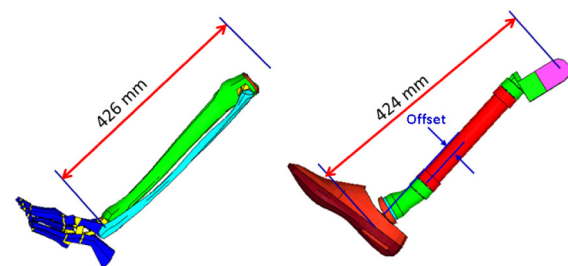


Figure 20. FHBM and HIII50 leg comparison

The pelvis kinematics of the two surrogates were also different in both restraint systems. Figures 21 and 22 show the x-y and x-z trajectories of the pelvis for the two occupants in the 35 mph ΔV PBLE pulse for belt-and-airbag and belt-only cases, respectively. The figures show that the differences between the pelvis trajectories were similar in both restraint systems for both occupants. The figures also show that both pelvises moved forward about 130 mm. The HIII50 moved to the left about 25 mm whereas, the FHBM moved to the right around 7 mm. Finally, the pelvis of the HIII50 moved down before it moved up whereas the FHBM pelvis moved only down and later in the event.

Tibia Transfer Equations

Three transfer equations were derived for each tibia load cell: axial force, x-moment, and y-moment. For the left upper tibia of the 55 YO FHBM F_z , M_x , and M_y were 75, 81, and 56% lower than those of the HIII50, respectively. The same responses of the right upper tibia of the 55 YO FHBM were 66, 76, and 51% lower than those of the HIII50, respectively. The 55 YO FHBM F_z , M_x , and M_y of the left lower tibia, were 80, 52, and 50% lower than those of the HIII50, respectively. Those values of right lower tibia of the 55 YO FHBM were 76, 36, and 48% lower than those of the HIII50, respectively (see Figures A9, A10, A11, A12, A13, A14).

Although the legs of the two surrogates are similar in length, there are several differences. First, there is no representation of the fibula bone and the Achilles tendon in the HIII50 leg. Second, there is a un-humanlike offset of the lower end of the leg (see Figure 20) that can produce moment during an axial loading of the leg. And third, the HIII50 leg is a rigid component compared to the deformable bones of the human leg.

Foot Transfer Equation

The 55 YO FHBM left and right foot accelerations were 14% lower than those of the HIII50. The feet of the HIII50 are rigid compared to human feet. In addition, the human foot has some cartilage between those bones that allow some flexibility in the human foot that is not represented in the HIII50 foot. There are no reported foot accelerations of the two surrogates under the same loading conditions (see Figure A15).

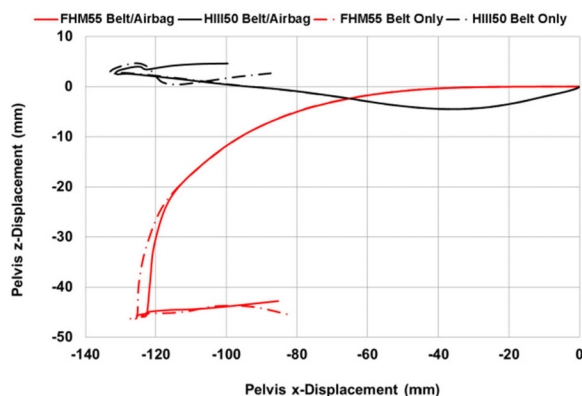


Figure 22. 55 YO FHBM and HIII50 pelvis trajectories 35 mph ΔV PBLE (z-x)

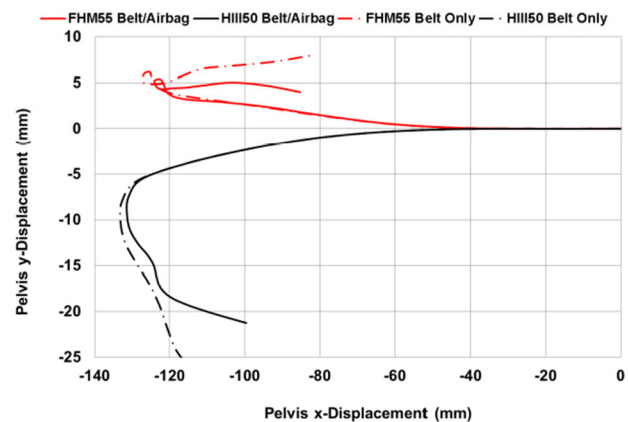


Figure 21. 55 YO FHBM and HIII50 pelvis trajectories 35 mph ΔV PBLE (y-x)

Shoulder Belt Force Transfer Equation

The two occupants were restrained with standard 3-point belt (no load limiter or pre-tensioner). Model-based transfer equation for the peak shoulder belt force was derived from the matched-condition simulations. The 55 YO FHBM shoulder belt forces were 12% higher than those of the HIII50 (see Figure A16). The test-based PMHS shoulder belt forces were 9% lower than those of the HIII50. Since the masses of the two surrogates are similar and they are subjected to the same pulse; it is expected that the belt forces would be comparable.

Model- and Test-Based Transfer Equations

There were limited published test data of the responses from matched-condition tests of the two surrogates. Similar to the model-based data, the same regression analyses were performed on the test-based data. Linear fit that passes through the origin was assumed in both the model- and test-based matched data. As seen from Figures 7, 8, 11, 12 A6, A7 and A16 and Table A5, the model-based slopes of the linear fits are consistent with those of the test-based transfer equations.

There is a need to generate additional test-based responses of the two surrogates under identical loading conditions. Different loading conditions, postures, and severities should be investigated. The range of those responses should be wide enough to include both the low and high stimuli observed in vehicular crashes.

HIII50-Based Head Injury Risk Curve

To give an example of how the transfer equations can be used; the skull fracture injury risk curve of Mertz, Prasad, and Irwin 1997 was transformed into the HIII50 domain using the HIC 15 transfer equation derived in the current study. The original risk curve was derived from cadaveric test data. The data consisted of isolated head drops on different surfaces, whole cadaveric sled tests against windshields, and helmeted cadaver drop tests. More information can be found in Prasad and Mertz 1985.