

Development of the MADYMO Race Car Driver Model for Frontal Impact Simulation and Thoracolumbar Spine Injury Prediction in Indianapolis-type Racing Car Drivers

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ABSTRACT

This paper describes the results of a project to develop a MADYMO occupant model for predicting thoracolumbar (TL) spine injuries during frontal impacts in the Indianapolis-type racing car (ITRC) environment and to study the effect of seat back angle, shoulder belt mounting location, leg hump, and spinal curvature on the thoracolumbar region. The newly developed MADYMO Race Car Driver Model (RCDM) is based on the Hybrid III, 50th percentile male model, but it has a multi-segmented spine adapted from the MADYMO Human Facet Model (HFM) that allows it to predict occupant kinematics and intervertebral loads and moments along the entire spinal column. Numerous simulations were run using the crash pulses from seven real-world impact scenarios and a 70 G standardized crash pulse. Results were analyzed and compared to the real-world impacts and CART HANS[®] model simulations. The Race Car Driver Model showed good kinematic correlation with the Human Facet Model and Hybrid III, 50th percentile male model. In addition, the RCDM predicted lower thoracolumbar loads and moments in the more upright, 25-degree recline position versus the normal, 45-degree recline position. Overall, simulations conducted in this study demonstrate the strong influence of spinal geometry and seating position on spine injury prediction outcomes during frontal impacts.

INTRODUCTION

Data taken from real-world Indianapolis-type racing car (ITRC) crashes indicate a high injury risk for the thoracolumbar (TL) spine in both frontal and rear impacts [1-3]. Trammell and Troxel *et al.* reported that the predominant injury type is a burst or wedge fracture resulting from an axial compressive load and forward flexion moment [1, 3]. Compared to TL spine injuries from rear impacts, the injuries sustained during frontal impacts are more severe, with an average severity index of 4.5 in the thoracic spine and 5 in the TL region using a scale from 1 to 6 defined by Trammell *et al* [1].

MADYMO Model Overview

Computer analysis programs, such as MADYMO (MAthematical Dynamic Model), have been used in the racing car environment to study crash simulations and the forces and accelerations acting on the occupant during these events [2-6]. The MADYMO Hybrid III, 50th percentile male dummy model is most frequently used to evaluate automotive safety restraint systems in frontal crash testing and is the standard representation of male race car drivers in the United States. Although the Hybrid III dummy represents the size and weight of the "average" U.S. adult male population, it lacks humanistic features such as a multi-segmented spine. To improve biofidelity and injury predictions of the Hybrid III dummy model, an ellipsoid spine was adapted from the MADYMO Human Facet Model (HFM) and integrated into the Hybrid III to form a new Race Car Driver Model

(RCDM). This integration allows for postural analysis and evaluations of the actual human body during frontal impacts in the ITRC.

In the MADYMO HFM, the vertebrae of the lumbar, thoracic and cervical spine regions are modeled as rigid bodies connected by free joints with lumped joint resistance (restraint) models [7]. In the lumbar and thoracic region, each vertebra is represented by a single ellipsoid. In the cervical region, each vertebra is extended with additional ellipsoids representing the transverse processes and spinal processes. The dynamic response of intervertebral discs, ligaments and the effects of muscular resistance are defined by non-linear lumped joint resistance models validated statically and dynamically with both post mortem human subjects (PMHS) and volunteer tests [7]. The neutral spine position of the HFM is represented by the curvature of an erect standing person (Figure 1).

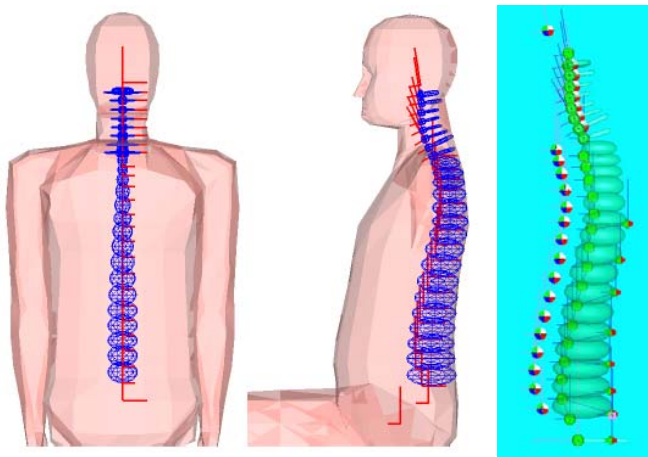


Figure 1. Neutral spine position of human facet model

Motorsports Spine Injury Research

To analyze the spinal contour of a seated ITRC driver, Trammell *et al.* used a modified Infinity Pro Series (IPS) Dallara chassis with a 50th percentile, male volunteer and x-rayed him in various orientations, including normal standing and seated posture as well as seated posture in the driver's own foam seat. The racing seat configuration used a 6-point restraint system, with and without the HANS®, in four configurations: upright (neutral racing position), inverted (head hanging), nose up (head back) and nose down (head forward). Results from this study showed that, in the racing seat configuration, the spine is pre-flexed forming a C-shape that eliminates the normal lordotic and kyphotic curvatures of the spine.

TL injuries from rearward crashes were investigated using various impact methods, including a full-scale crash test of an Indy Car (CART) racecar into a rigid barrier at 50 mph (80 kph) and numerous HYGES sled tests with a pulse magnitude of 60 G's [1]. For the full-scale crash test, an instrumented Hybrid III, 50th

percentile male anthropomorphic testing device (ATD), or dummy, was outfitted with a racing helmet and HANS® device to accurately represent the racing car driver. In the HYGES sled tests, the THOR ATD was instrumented with T8 and T12 load cells and seated in the sled buck using the reclined position of an ITRC driver. Trammell *et al.* reported that the measured lumbar compression loads in the Hybrid III during the barrier test were 10 times greater than the 6000 N accepted injury threshold, and that the chassis experienced an obvious buckling effect near the location of the accident data recorder (ADR), causing a spike in the z-axis acceleration data [1]. For the THOR ATD sled tests, results also showed high TL z-loads including 8,360 N at T8 and 8,430 N at T12. The authors concluded that the loads predicted by the Hybrid III and THOR were sufficient enough to produce vertebral fractures in the ITRC driver setting [1].

The CART HANS® MADYMO model developed by Weerapuli *et al.* consists of a Hybrid III, 50th percentile male dummy model in a finite element (FE) CART car [4-6]. Troxel *et al.* used this model to investigate five TL injury cases from frontal impacts and two non-injury cases [3]. The CART HANS® model predicted unusual kinematics and excessive vertical spine loading when the z-acceleration from the ADR was included as part of the model input. As a result, this loading component was subsequently removed from the model input based on the supporting information that: 1) the vehicle chassis buckles near the ADR mounting location, which can cause inaccurate acceleration readings along this axis [1]; and 2) the driver cannot be seen bouncing out of the seat in crash videos. Once the z-acceleration input was removed from the five injury case simulations, driver kinematics returned to normal and TL z-loads ranged from 4,825 N to 13,211 N, while y-moments ranged from 146 Nm to 622 Nm. The authors concluded that the MADYMO Hybrid III, 50th percentile male dummy model overestimates TL loads. In addition, because the model has a rigid spine, it is unable to predict loads and moments in more than one location in the TL region. The conclusions from this study suggested a need to use a validated, biofidelic mathematical model with a multi-segmented vertebral column to predict TL spine injury potential in ITRC drivers [3].

Trammell *et al.* used the MADYMO THOR dummy model and the Human Finite Element Model (HFEM) to evaluate rear impacts in IndyCars and predict spinal injuries in racing car drivers [2]. The MADYMO THOR dummy model was used to compare simulation output with sled test results to validate the vehicle environment model and the interaction between the vehicle and the occupant models. After the vehicle model was validated, the THOR dummy model was replaced with the MADYMO HFEM. This newer model version has a multi-segmented finite element spine, but it was not designed for severe impacts nor is it easily positioned in the reclined ITRC driver position. Extensive measures were taken to position the HFEM in the 45-degree reclined orientation of an ITRC driver, but the orientation

was not optimal. One positioning method rotated the HFEM as if it were a rigid body, but this placed the legs up in the air, the arms reaching vertically and the head and eyes angled approximately 45-degrees upward. A pre-positioned HFEM from TNO, the manufacturer of MADYMO, placed the occupant's legs on the floor, but the head and eyes were angled 45-degrees upward rather than horizontally and the FE spine had unrealistic initial deformations [2]. Once the HFEM was positioned in the two best possible orientations given the ITRC environment, a 6-point FE belt restraint system was added along with a simplified ellipsoid HANS® and helmet model. A severe, 70 G rearward crash pulse was used as initial input while the occupant was oriented in the above two configurations. Trammell *et al.* reported that the rotated HFEM predicted realistic kinematics with high spinal compression loading in the z-direction, but the pre-positioned HFEM produced unrealistic kinematics and became unstable 35 ms into the simulation [2]. The authors concluded that the MADYMO HFEM is not suitable for the study of severe rear impacts in the Indy Car setting and, as an alternative, suggested using the MADYMO HFM. The results of the HFM simulations are briefly reported on by Trammell *et al.* and state that the model reliably and reproducibly predicted large spinal compressive loads in the z-direction [1].

Considering the recurring problems associated with simulating ITRC driver impacts using MADYMO dummy and human models that are designed for upright seating positions, this project proposed to develop a new MADYMO RCDM for the primary purpose of simulating frontal impacts in the ITRC environment to predict intervertebral loads and moments along the spinal column. This paper discusses the MADYMO RCDM development and analyzes the ability of the RCDM to predict occupant kinematics and intervertebral loads and moments in the TL spine region. Potential factors influencing spinal loads and moments in ITRC drivers during frontal impacts were also analyzed. These include spinal curvature, seat back angle, shoulder belt mounting locations, and under thigh seat pans referred to as a leg hump.

METHODS

Initial Modeling using the MADYMO HFM

Preliminary frontal impact simulations were run using the 50th percentile male MADYMO HFM in the same car model as used by Trammell *et al.* [2]. Kinematic analysis of the HFM showed the driver moving forward, contacting the belts, followed by complete distortion and disruption of the occupant model at 22 milliseconds (Appendix A). Despite modifications to model contacts between ellipsoids, car structures and restraints, the same kinematic response was reproduced in other frontal impact cases using the actual crash pulses reported by Troxel *et al.* [3]. It was concluded that the current HFM could not be appropriately applied for the

study of injury responses to ITRC drivers during frontal impacts and therefore, a new model was created using appropriate components of both the MADYMO Hybrid III, 50th percentile male dummy model and the ellipsoid spine from the HFM.

Race Car Driver Model Development

The RCDM was developed in XMG 5.1 using MADYMO version 6.4 by integrating the HFM ellipsoid spine with the ellipsoid Hybrid III, 50th percentile male, dummy model. Modifications were made to the ellipsoid chest, sternum, ribs and abdomen to accommodate the multi-segmented spine. Figure 2 shows the neutral posture of the driver model in a normal, upright seating position. In the neutral position, the spinal curvatures were biofidelic in that the thoracic spine was kyphotic while the cervical and lumbar spine was lordotic. The sacral region of the spine was not represented as a multi-segmented portion because it is fused to the pelvis and, thus, the sacrum and pelvis were represented as a combined feature in a large ellipsoid.

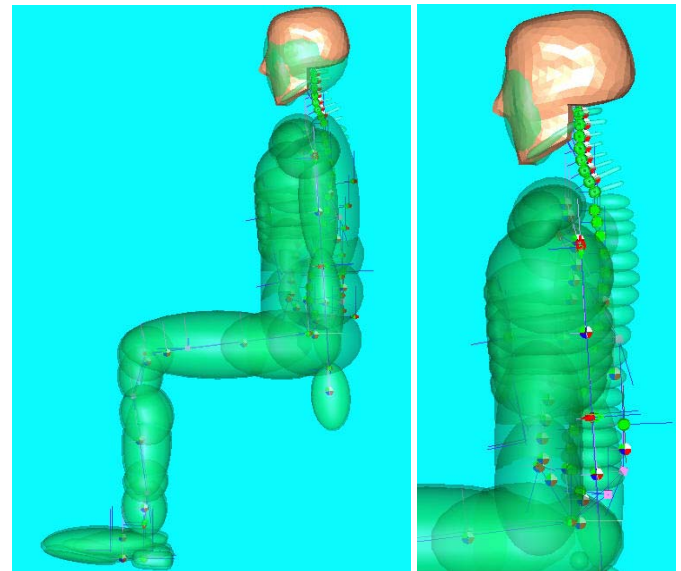


Figure 2. RCDM in neutral position

Initial positions of the ellipsoid spine segments were modified in the MADYMO include file of the main input deck. The change in spinal curvature was modeled based on research performed by Trammell *et al.* in which lateral x-rays of an actual race car driver showed the loss of normal spinal curvatures in both the thoracic and lumbar spine when positioned in a reclined driving position [1].

Validation

Following development of the RCDM, validation was performed in two seating configurations: an upright seating position and a 45-degree reclined seating position. First, the RCDM kinematics were compared to those of the Hybrid III, 50th percentile male dummy

ellipsoid model using an application file supplied by MADYMO (Filename: e_hyb350fc.xml). This application file was chosen because it is a standard frontal impact simulation using a 3-point finite element belt system and a 90-degree seat reclined at a 10-degree angle. The application file was originally developed by MADYMO for the Hybrid III, 50th percentile male dummy facet model and was subsequently replaced by the Hybrid III, 50th percentile male dummy ellipsoid model and the RCDM. Appendix B shows 20 millisecond lateral views of the simulation with the RCDM on the left and the Hybrid III ellipsoid model on the right. Overall, the RCDM kinematics closely resembled the Hybrid III dummy kinematics with the exception of the spine, which was more flexible in the RCDM. In particular, the RCDM's neck can be seen to flex more than the Hybrid III ellipsoid model from 120 ms to 180 ms. This is to be expected, as the Hybrid III model spine is rigid and it does not flex like that of a more biofidelic, multi-segmented spine.

The second form of validation compared the RCDM kinematics to actual crash video from the seven crash cases (5 injury, 2 non-injury cases) previously reported by Troxel *et al* [3]. The helmet and 6-point restraint system from the CART HANS[®] model were upgraded to MADYMO version 6.4 and adapted to fit the driver (Figures 3 and 4). No head and neck support device was used for these simulations because there is no record of drivers wearing such devices at the time these crashes occurred.

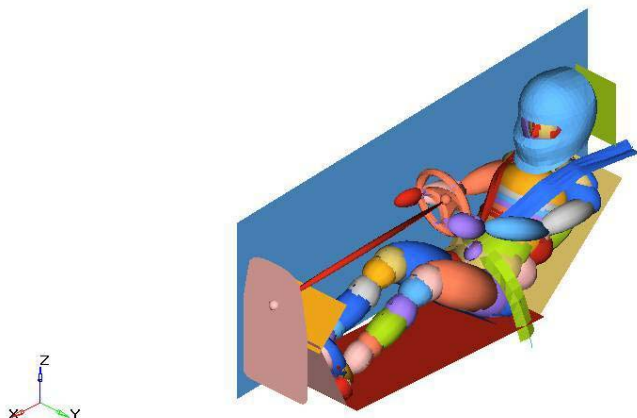


Figure 3. RCDM positioned in cockpit with left side panel removed

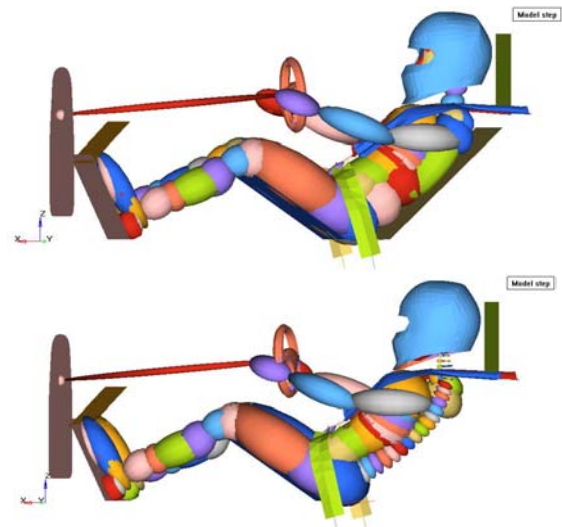


Figure 4. Side views of RCDM seated with C-shaped spinal curvature geometry

A rigid plane cockpit was designed to represent a simplified seating compartment, and a steering wheel was adapted from the CART HANS[®] model. The seat was reclined 45-degrees from vertical, and a planar leg hump was implemented underneath the driver's thighs. The driver model settled into the seat under gravity and the belts were positioned using the BELT_PRETENSIONER.PAYIN_TIME option. Pretensioners were removed prior to running case simulations. As described above, positioning of the driver in the race car seat oriented the ellipsoid spine in a way that eliminated the kyphotic curvature in the thoracic spine and the lordotic curvature in the lumbar spine.

The crash pulses used to kinematically validate the RCDM in the 45-degree reclined seating position were selected from those associated with the seven real world crash cases previously reported by Troxel *et al* [3]. Model kinematics were validated using case B video footage, which showed the driver's motion in the cockpit at the time of the impact and did not show the driver bouncing up and out of the seat. Analysis of the video captured at 24 frames per second (fps) in one of the selected cases (Driver B) showed the car contacting the concrete barrier on the outside of an oval track with the front nose of the car. The driver can be seen moving forward until his helmet contacted the steering wheel and surrounding interior car structures. Following the frontal impact, the driver appeared to settle back down into the seat.

When only x- and y- accelerations were included in the crash pulse, the motion of the RCDM reasonably matched the driver's motion seen in case B (Appendix C). Therefore, based on the artifacts previously reported by Troxel *et al.*, z-accelerations were removed from all crash pulses applied to the RCDM [3]. At the end of each simulation, the driver model settled back into the seat. Model validation was concluded as insufficient for more precise validation. **Documents**

detailed simulations and analysis of the seven crash cases were prepared accordingly.

Real World Frontal Impacts

Eight crashes were simulated using the above described RCDM and rigid plane seating configuration, which most closely mimicked the actual seating configuration. The RCDM's spine was positioned in the shape of a "C" to represent that which was observed in driver x-rays. Results were analyzed using Altair Hyperview 7.0 Animation and Plot features. Spinal load and moment outputs were analyzed from the sacrum to T4 giving a range from four levels below the thoracolumbar region to five levels above the highest injury level. Loads and moments were measured at the anterior portion of the vertebral ellipsoid where the bracket joints and rigid bodies are located.

Parametric Study

Seating and belt configuration was hypothesized to have a significant effect on the loads and moments experienced by the TL spine during frontal impact. Therefore, a parametric study was designed based on the use of a standard crash pulse and results from a report by Melvin *et al.* on racing car restraint system performance during frontal crash testing in Indy cars [8]. Three seat back angles were chosen based on the traditional 45-degree recline position and the more upright passenger car seating position (25-degrees), which allows for less restraint loading on the human body, thereby reducing spinal loads. Three shoulder belt anchoring locations were chosen based on Melvin *et al.* findings (1994) that raising the shoulder belt anchors until they are even with the top surface of the shoulders reduces neck tension load and provides better initial restraint of the torso when compared to the lower anchor location (75 mm below horizontal) [8].

A standard crash pulse used for ITRC sled testing is a 70 G (686.7 m/s²) peak acceleration designed by the SFI Foundation, Inc. The SFI Foundation, Inc. is a non-profit organization established to issue and implement standards for specialty/performance automotive and racing equipment. The SFI pulse represents a severe frontal impact (x-acceleration) and was used as the primary model input for the parametric study (Figure 5).

Simulations were run with and without the leg hump using a range of recline angles (measured from vertical) to represent the standard ITRC driver position (45°), a position half way in between the racing car driver and passenger car driver seating position (35°), and the passenger car driver position (25°). In each of these configurations, shoulder belts were anchored horizontal to the shoulders (0 mm), 75 mm above the shoulders (+75 mm) and 75 mm below the shoulders (-75 mm). Table 1 summarizes the 18 simulations performed during the parametric study and Appendix E shows the

initial position model set-up for each case. Positioning of the RCDM spine was easily achieved by changing the initial positions of the vertebral ellipsoids in the MADYMO main input deck. The C-shaped spine was modeled through variations in these parameters.

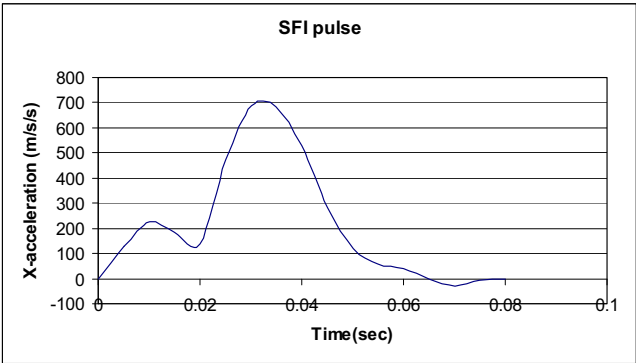


Figure 5. SFI Crash Pulse

Parametric Study			
	25° recline With leg hump	35° recline With leg hump	45° recline With leg hump
Shoulder belt anchor points	75 mm above	75 mm above	75 mm above
	At shoulder level	At shoulder level	At shoulder level
	75 mm below	75 mm below	75 mm below
	25° recline Without leg hump	35° recline Without leg hump	45° recline Without leg hump
Shoulder belt anchor points	75 mm above	75 mm above	75 mm above
	At shoulder level	At shoulder level	At shoulder level
	75 mm below	75 mm below	75 mm below

Table 1. Parametric Study Outline

Spinal Curvature

The effect of spinal curvature on vertebral body load and moment is of interest as it may influence racing seat design. The validated RCDM was therefore used to investigate two variations in spinal curvature. Assessments were made based on the analysis of data generated during the parametric study (see Results, below), and it was concluded that a 25-degree reclined seat back with shoulder belts anchored 75 mm above horizontal and the presence of a leg hump underneath the driver's upper legs would be the optimal seating configuration to reduce TL spine fractures during frontal impacts. Crash pulses from the real-world cases were used as the primary model input, and simulations were run using MADYMO version 6.4 and the RCDM. Two spinal column geometries were simulated using the proposed optimal seating configuration. The first geometry was the C-shaped curvature preserved from

the parametric study. The second geometry was the natural curvature consisting of a kyphotic thoracic spine and a lordotic cervical and lumbar spine. Figure 6 shows the initial conditions of the driver model in the upright, 25-degree seating configuration with both a C-shaped spine and the natural spinal geometry. Initial positions of the ellipsoid spine segments were modified in the MADYMO main input deck to achieve the C-shaped spine and natural geometry.

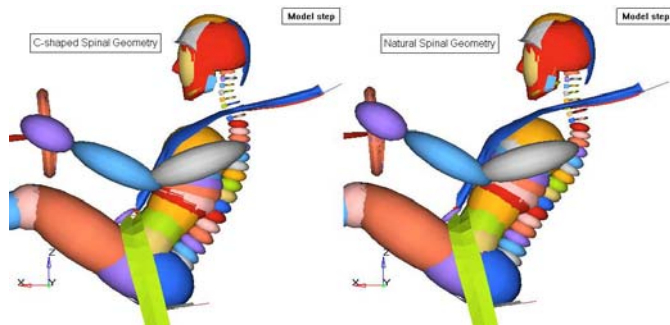


Figure 6. C-shaped vs. natural spinal geometry

Potential Injury Criteria

The maximum compressive load and maximum flexion moment predicted by the RCDM in the analysis of the real world cases did not appear to adequately differentiate between injury and non-injury cases. Therefore, two other possible prediction parameters were also investigated using the current, common ITRC seating configuration.

To evaluate the influence of combined compression and flexion on TL injury outcome, the neck injury criteria (Nij) was adapted to yield a thoracolumbar injury criterion (TLij). A neck injury criterion (Nij) is a measurement used to predict injury likelihood based on critical limits for neck axial loading and bending moments. Nij is defined by:

$$Nij = \frac{Fz}{Fz_{critical}} + \frac{My}{My_{critical}}$$

Fz = axial load
Fz_{critical} = critical load value
My = bending moment
My_{critical} = critical bending moment

Injury is anticipated for Nij values greater than 1. This parameter has been used to monitor and prevent critical neck loads via a linear combination of tension/compression and flexion/extension moments.

The TLij was defined to parallel the calculation of the Nij:

$$TLij = \frac{Fz}{Fz_{critical}} + \frac{My}{My_{critical}}$$

Fz = axial load
Fz_{critical} = critical load value (10,000 N)
My = bending moment
My_{critical} = critical bending moment (-300 Nm)

For the TL region, the maximum axial compression value (Fz_{critical}) was defined as 10,000 N and the maximum flexion moment (My_{critical}) was defined as -300 Nm. These values were selected based on the maximum values predicted using the MADYMO Hybrid III, 50th percentile male dummy model and RCDM and then narrowing down the value to a critical limit that would most likely cause injury in the TL region.

Duration of loading has also been hypothesized to be an important factor in the occurrence of injury. In order to evaluate the duration of peak compression loading in the 45-degree reclined position with leg hump and horizontal shoulder belts, half of the peak load was calculated from L1 to T9 and the width of the peak at half the load was measured to assess the time duration. It was hypothesized that the likelihood of injury will increase with longer force durations, whereas the risk of injury will decrease with shorter durations.

RESULTS

Real World Frontal Impacts

The model predicted realistic driver kinematics for each case, with more spinal flexibility than that of the Hybrid III, 50th percentile male dummy model. Unusual neck flexibility and elongation were observed in severe cases but were not predicted in less severe crashes. Neck kinematics will be discussed further in the discussion section. Appendix D shows two tables summarizing peak spinal loads in the z-direction from the sacrum to T4 and peak spinal moments about the y-axis from the sacrum to T4. In both tables, the highlighted cells represent the level at which the driver was injured, from L1 to T9.

Parametric Study

Eighteen cases were simulated using the SFI crash pulse and RCDM in various seating orientations. Results were analyzed using Altair Hyperview 7.0 Animation and Plot features. Overall, the model predicted realistic driver kinematics but neck flexibility kinematics continued to look unusual. When the shoulder belts were anchored at 75 mm below horizontal, they slid off of the shoulders laterally. With a leg hump and the seatback reclined to 25 degrees, the driver slid up the leg hump before being restrained by

the lap belt. At 25-degrees, the driver flexed forward about the pelvis and was quickly restrained by the lap belt. When the leg hump was removed, the RCDM slid further forward along the bottom of the seat structure, and the foot pan was the primary structure preventing forward movement. During simulations, removal of the leg hump also allowed the driver's knees to contact the underside of the steering column.

Loads and moments were measured at the anterior portion of the vertebral ellipsoid where the bracket joints and rigid bodies were located. These simulations show high lumbar compression loads beginning at the sacrum and continuing up the spine to L1. Peak z-loads and moments about the y-axis for the thoracolumbar region are shown in Appendix F. In the thoracolumbar region (L1-T9), the highest compression loads are observed at the lowest level of L1. Compared to the baseline 45-degree recline with leg hump and horizontal shoulder belts, compression loading decreased at L1 when the shoulder belts were anchored 75 mm above horizontal. When the leg hump was removed, compression loads increased at L1 but decreased at T10. High thoracolumbar moments were observed at T10 for all conditions, while the lowest moments were predicted at L1. The majority of thoracolumbar spine moments were reported as negative values, indicating that the spine was in flexion during the peak moment. Moments tend to increase when the leg hump is removed. Overall, compression and flexion measurements decreased at a 25-degree seat angle when compared to those predicted in the 45-degree recline position when a leg hump was included and belts were mounted above the horizontal (Figure 7). While it resulted in only a slight decrease at most levels, this was deemed to be the optimal seating configuration to minimize spinal loading.

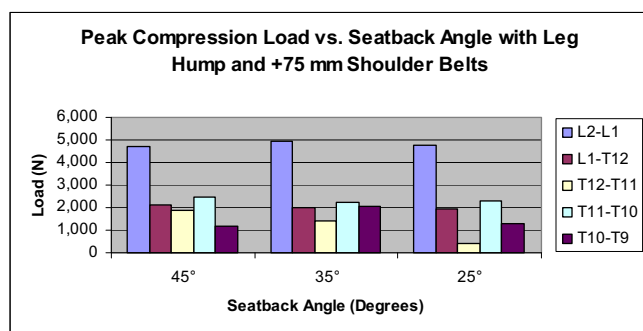


Figure 7. Peak loads vs. seatback angle

There was significant variation observed in peak compression and tension loads between adjacent TL vertebrae in some cases. In these simulations, where there was high compression loading at one level followed by low compression loads at the next level, the spine was taking on its natural curvatures of lordosis in the lumbar region and kyphosis in the thoracic region (Figures 8 and 9). This explains why certain vertebrae were in tension while others were in compression.

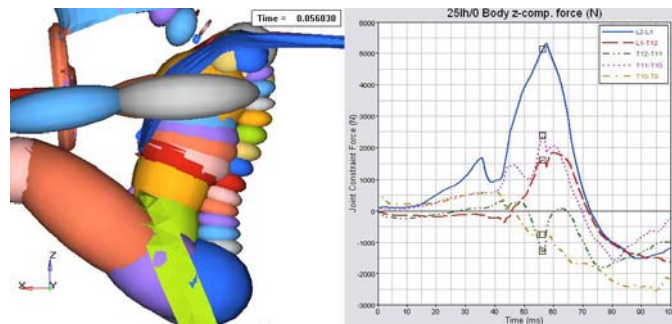


Figure 8. Compression and tension loads in 25lh/0

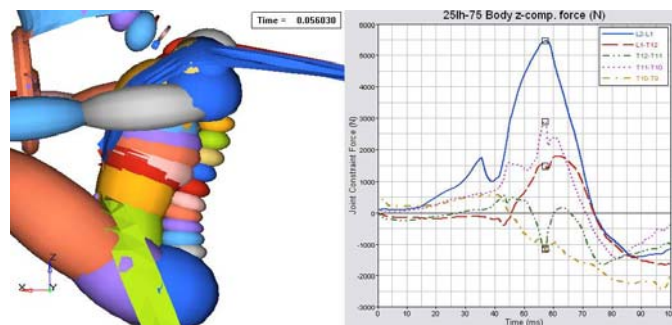


Figure 9. Compression and tension loads in 25lh-75

Spinal Curvature

The eight crash cases were simulated using two spinal geometries in a 25-degree reclined seat back with a leg hump and shoulder belts mounted 75 mm above horizontal. Results were analyzed using Altair Hyperview 7.0 Animation and Plot features. Once again, the model predicted realistic driver kinematics for each case and neck elongation was minimized when the RCDM's spine maintained a natural curvature. The C-shaped spinal geometry caused the helmet to be positioned higher up on the head so that the helmet bottom did not contact the chest. When the helmet was lower on the head, as in the simulations with a natural spinal curvature, the neck did not stretch as much as in the C-shaped spinal geometry simulations. Appendix G shows 20 millisecond clips of case A1 with the C-shaped spinal geometry on the left and the natural spinal geometry on the right. The thoracic kyphosis and lumbar lordosis are clearly visible in the natural spinal geometry and were maintained until the driver's head contacts the steering wheel.

Compression and flexion outputs were analyzed from the sacrum to T4 for each impact case and spinal geometry. Peak compression loads and flexion moments for the thoracolumbar region are shown in Appendix H. Loads and moments were measured at the anterior portion of the vertebral ellipsoid where the bracket joints and rigid bodies were located. In general, the natural spinal geometry produced lower compression loads but higher flexion moments compared to the C-shaped spinal geometry. These results suggest that