PARAMETRIC ANALYSIS OF VEHICLE DESIGN INFLUENCE ON THE FOUR PHASES OF WHIPLASH MOTION

Polat Sendur, PhD; Robert Thibodeau; John Burge – Crashport
Allan Tencer, PhD – University of Washington

ABSTRACT

The occupant motion in rear impacts that is associated with Whiplash Associated Disorders (WAD) can be broken into four phases, each representing a distinct possible source for injury: retraction, extension, rebound and protraction. Injury criteria such as NIC and N_{km} have been proposed to evaluate the injury potential of these mechanisms. Key vehicle design factors contributing to WAD motion include head restraint backset and height, seat stiffness, and vehicle structural characteristics. The hypothesis of this study was that different WAD phases would show different levels of sensitivity to vehicle design factors. A detailed 50th percentile male model with a biofidelic neck was used in a 100-run Monte Carlo analysis of a rear impact, varying design factors across the values documented in literature. Total energy was held constant and Delta V was 10 kph. Vehicle stiffness had strong influence on the retraction (70%), rebound (43%) and protraction (47%) phases. Backset (the initial distance from occupant head to vehicle headrest) demonstrated a strong influence on the extension (49%) and rebound (39%) phases. For design rating and WAD protection, it is important to view the vehicle as a system and not just a single design variable.

Keywords: Cervical Spine, Whiplash, Mathematical Models, Multi-body Dynamics, Sensitivity Analysis

WHIPLASH ASSOCIATED DISORDERS (WAD) continue to be the most frequent injuries resulting from rear-end motor vehicle accidents. As a result, attention is being placed on how vehicle design features can reduce these types of injury. While the precise injury mechanisms of WAD are not yet fully understood, it is well established that injury can result from sudden differential motion between the head and torso.

Testing vehicle components for WAD injury potential is expected to influence future vehicle design decisions through publicly-disclosed ratings similar to those provided for high-impact crash protection [22]. The design and rating of vehicles regarding whiplash injury protection may prove challenging due to multiple design factors that could affect the underlying motions: head restraint position [10, 44, 51], head restraint stiffness [7, 12, 16], seat energy-absorption characteristics [16, 32, 35, 46, 50] and vehicle structural characteristics [23, 40, 51].

OCCUPANT MOTION IN REAR IMPACTS: The occupant motion occurring during rear impacts can be broken into multiple phases: retraction, extension, rebound and protraction. Each phase represents a different potential mechanism or set of mechanisms, receiving varying levels of support from the biomechanics community as being responsible for whiplash injuries [32, 50]. Injury criteria have been proposed for each phase. In addition, injury thresholds have been proposed and evaluated for some of these mechanisms.

RETRACTION PHASE: In the retraction phase, the upper torso is pushed forward by the seat back while the occupant's head remains essentially stationary. In this phase, the cervical spine has been shown to take on an S-shape by forcing the upper cervical spine into flexion and the lower cervical spine into extension [12]. The retraction phase ends when maximal retraction is reached, which is often caused by head restraint contact. A primary injury mechanism believed to be associated with this motion is a pressure spike caused in the spinal canal by the rapid relative linear

motion of C1 with respect to T1 during retraction [2, 3, 46]. The Neck Injury Criterion (NIC) was explicitly developed to quantify the risk for WAD based on this mechanism [3]. Several definitions of NIC are in use by researchers [3, 8, 13]. In this paper we used the definition used in [13]. NIC values greater than 15 m^2/s^2 are believed to be associated with the potential for long-term symptoms [3, 37, 51], although some believe this may be high [8]. A NIC reading of 12 m^2/s^2 appears to be about as high as researchers are willing to ethically push human subjects [38, 50], although a few exceptions up to 19.8 m^2/s^2 are documented in the literature [50].

EXTENSION PHASE: This phase can occur when the head reaches a point of maximal retraction before striking the head restraint, causing the occupant's head to rotate rearward, which causes upper cervical motion segments to join the lower motion segments in extension orientations. It can also occur when there is no headrest, the headrest is set too low, or there is significant upward sliding of the occupant, each resulting in the head and neck extending backward over the top of the seat-headrest. The primary injury mechanism associated with this motion is hyperextension. This mechanism was the first cited for WAD causation as early as the 1950's [24, 25] and has often been measured by the extension moment exerted at the atlanto-occipital joint or by whether normal range of motion has been exceeded. Extension values exceeding 30.5 Nm are believed to present significant potential for injury based primarily on the work of Mertz and Patrick [29, 30]. The Society of Automotive Engineering (SAE) Standard J885 incorporates this limit into its guidelines for human subject testing [41]. Only one set of volunteer tests was found in the literature that approached or exceeded this limit [29].

REBOUND PHASE: The rebound phase can occur when the occupant's head contacts the head restraint or the seat reaches maximum elastic rotation and returns the spring energy into the occupant's torso and head. Head restraint-seat rebound causes the highest head translational accelerations to occur as well as peak axial and shear forces. The cervical spine is known to be particularly vulnerable to injury when significant torque moments are combined with significant shear or axial forces. The N_{ij} criterion is currently used by the National Highway Transportation Safety Administration (NHTSA) to assess neck injury potential in higher impact crashes by combining neck axial forces and torques [17]. In a similar manner, the N_{km} criterion has been proposed to evaluate the effect of combined shear forces and moments [32] for assessing the potential for WAD injury. An N_{km} value of 1.0 has been proposed as a threshold for AIS 1 neck injuries, although other studies suggest the threshold could be lower [21].

PROTRACTION PHASE: The protraction phase can occur after rebound when differential motion between the head and torso are reversed. This phase can become significant when forward motion of the upper torso is arrested by the seat belt shoulder strap. The mechanisms associated with this motion are similar to those in the retraction and extension phases when the cervical spine transitions from an S-shape toward full flexion. A pressure spike in the spinal canal has also been proposed as a potential mechanism for whiplash injuries occurring in frontal impacts, which could also be measured by the NIC [3]. Moments, axial, and shear forces are also presented by forward motion of the head relative to an upper torso restrained by a shoulder harness. Difficulties in deriving these values from rear impact ATDs have led many researchers to look at head and T1 rebound velocities to assess rebound and protraction-phase effects [32].

INFLUENCE OF DESIGN FACTORS ON WAD PHASES: The contribution of each design factor to the distinct phases of the WAD motion has not been fully explored. The current study evaluates the sensitivity of the proposed WAD injury criteria associated with each phase to variations in key vehicle design features. It applies a parametric analysis approach to investigate these relationships using a detailed mathematical human model.

MATERIALS AND METHODS

HYPOTHESIS AND TEST SETUP: The hypothesis tested in this study was that the different whiplash phases would show significantly different levels of sensitivity to variations of the key vehicle design factors. A combined mathematical vehicle and occupant model was used to test this hypothesis for a well-aligned rear-impact crash. The mathematical modeling employed a

representation of the vehicle bumper system and occupant and a model of the seat. A 100-run Monte Carlo analysis was conducted, varying key vehicle design factors across their expected range of real-world values based on existing literature. Total energy was held constant and all runs were carried out at a Delta V of 10 kph. The influence of the design factors was assessed using the proposed injury criteria associated with each phase of the WAD motion.

DESCRIPTION OF THE MODELING ENVIRONMENT: Multi-body models were constructed in MSC.ADAMS [42, 43], a general purpose simulation environment widely used in automobile industry. The occupant was modeled as a 50th percentile male, incorporating individual cervical and lumbar vertebrae. The base model is comprised of 24 segments and 23 joints (head, cervical spine, torso, lumbar spine, pelvis, upper and lower arms, upper and lower legs, and feet). Occupant component motions are based on appropriate component mass properties, dimensions, joint stiffness and damping friction and limits obtained from various published sources. The basic layout of the parts, joints and forces in the model can be found in Sendur, et al. [43].

The human cervical spine model is based on the De Jager model [11], and is similar to that used on other research studies [20]. The upper spine model includes seven vertebrae (C1-C7), the head and the first thoracic vertebra (T1). The rigid bodies in the upper spine model are connected through two-axis nonlinear viscoelastic joints, which represent the mechanical behavior of the intervertebral discs, ligaments, facet joints and muscles. The neck joints are permitted movement in the sagittal and lateral planes but not about the spinal axis. The head is permitted motion about all three axes. The force-deflection relationships used in the model are nonlinear, and the moments are non symmetric for motions about any axis.

A similar modeling strategy is taken for modeling the lumbar spine. The lumbar spine is comprised of five vertebrae, L1 through L5. These rigid bodies are connected by two-axis viscoelastic elements, which represent the mechanical characteristics of the inter-vertebral discs, ligaments and muscles. The kinetic characteristics of the lumbar spine were calibrated to the human test data available in the literature by iterating the model until the best match was obtained [1]. The thoracic spine was modeled as a single segment given the restriction in movement provided by connections to the rib cage.

The seat model is comparable to other models published in the literature [13, 50]. The seat model is comprised of three bodies - a seat bottom, seat back, and a head restraint. The seat back is oriented with respect to seat bottom at a specified angle, and is constrained with a hinge joint. The seat characteristics represented include the seat back hinge stiffness, the stiffness of the seat cushion (between various body parts and the seat including pelvis-seat bottom, pelvis-seat back, lumbar spine-seat back, torso-seat back), and the friction coefficients between occupant body parts and seat. A coefficient of friction of 0.7 was used for contact between the occupant and the seat, with the exception of the head contact with the head restraint, which was modeled using a coefficient of friction of 0.5 [18]. The head restraint is modeled as a deformable body fixed to the seat back with a specified backset (horizontal distance between the back of the head and the front of the head restraint) and headrest height [18]. Lap and shoulder belts are modeled with nonlinear force deflection characteristics. The protraction phase may become significant when forward motion of the upper torso is arrested by the shoulder strap. The kinematic responses of the occupant model to crash motions were compared with the responses of human volunteers subjected to sled accelerations simulating rear impacts at a Delta V of 9 km/h [5]. The comparison dataset consisted of 43 tests using 19 human subjects, and Hybrid III and RID2 ATDs. Details of the comparison methodology and results are found in the Appendix A.

VARIATION IN VEHICLE STRUCTURAL FACTORS: Vehicle acceleration history or crashpulse has been shown to vary significantly in both shape and duration in impacts of similar velocities [23, 51]; this variation translates into variations in occupant motion [5, 6, 10, 11, 13, 19, 20, 23, 43]. An important contributor to the crashpulse that results from a rear-end impact is the force-deflection characteristics of the two colliding vehicles, which can vary widely depending on the

Table 1: Parameter Ranges

Factor	Avg.	Std. Dev.
Crash Duration (ms)	105	75
HR backset (cm)	8.9	5.8
HR height (cm)	8.4	4.6
Seat hinge (N*m/deg)	60	25

vehicle design. For this study, various combinations of vehicle force-deflection curves, derived from the literature [14, 15] or from crash testing [33, 43], were used to determine expected real-world durations for 10 kph velocity changes. The resulting crashpulses represented highly complex shapes that did not lend themselves to the requirements of this study for understanding injury sensitivities at a high level. Idealized pulses (single peak, sine wave) were created using the resulting durations, which ranged from 30 ms to 180 ms with peak accelerations ranging from 2.5 g's to 14.9 g's (Figure 9 in Appendix B). A normal distribution cutoff at the first standard deviation was used to grossly approximate the real world distribution of impact durations between 30 ms and 180 ms (see Table 1). Although the pulse shapes used in this study are idealized, they can be expected to represent the minimum complexity of the relationships between the vehicle design factors and WAD injury mechanisms.

VARIATION IN SEAT FACTORS: Seat geometry is another aspect of vehicle design known to influence occupant motion. In particular, the height and backset of the head restraint have been identified as key design factors affecting injury by affecting when and how the head is accelerated [10, 51, 44]. In this study, backset was varied from 3.1 cm to 14.7 cm using a normal distribution based on the observational study by Cullen, et al. [9]. Head restraint height was varied from 3.8 cm to 13 cm using a normal distribution also based on the same study. All head restraint positioning is based on the positioning system found in the Research Council on Automotive Repair (RCAR) standard and used by the Insurance Institute for Highway Safety [39]. Seat energy-absorption characteristics also affect occupant motion in rear impacts [16, 32, 35, 50]. Several studies have been conducted to characterize the rotational stiffness of different seat designs [45, 31, 49]. Because of the trend toward stiffer seats over the last several years, the values found in the recent work of Viano and Gargan [49] were used for seat stiffness variation using a normal distribution, ranging from 35 Nm/deg to 85 Nm/deg.

SENSITIVITY ANALYSIS: Four predictor variables related to automobile design parameters were selected based on published results and engineering judgment: crash duration, headrest backset, headrest height, and seat stiffness. Then a systematic search was undertaken to identify the variables that were the best predictors of the individual WAD effects for each WAD phase. Scatter plots were made for each injury measure and each predictor variable to get some notion of the most likely relationships and to see how well the single variable relationships behaved. Trial single-variable regressions were performed. When more than one predictor variable showed good correlation or a high regression coefficient, multivariable regression techniques were applied. The end result was a set of numbers quantifying the contribution of each predictor variable during each WAD phase.

RESULTS

Variation of selected injury measures is presented below. All head accelerations are measured at the center of gravity, while forces (shear and axial) and moments (extension and flexion) are measured at the Occipital Condyles (OC), which is also used as the reference point for the top of the cervical spine to derive extension and flexion moments.

Most of the runs resulted in contact between the head and head restraint. Head accelerations (shown in Figure 6) ranged from 4.3 g's to 14.2 g's. High head rotational accelerations corresponded with contact between head and head restraint. Head angular accelerations ranged from 73 rad/s2 to 1955 rad/s2, with a few runs exceeding 1800 rad/s^2, which is often used as a threshold for non-contact concussion [36]. NIC ranged from a low of 1.53 m^2/s^2 to a high of 21.22 m^2/s^2 with over a third of the results exceeding the proposed threshold of 15 m^2/s^2 [32]. NIC frequency is shown in Figure 1.

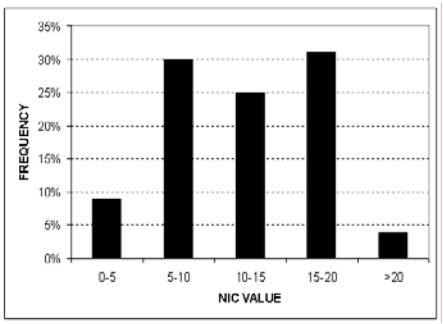


Figure 1: NIC Frequency Plot

Tension forces varied from 1 [N] to 500 [N] with 13 % of the runs exceeding the limits for human subject testing specified by SAE J885. Compressive forces varied from 39 [N] to 372 [N] with 9 % of the runs exceeding the SAE J885 recommendation [41]. Neck moments also varied from a low of 2.5 [Nm] to a high of 27.4 [Nm]. None exceeded the limits for human subject testing specified by SAE J885.

Table 2: Injury Measure Variation

Injury Measure	**Low**	**High**	**Avg.**
NIC (m2/s2)	1.53	21.22	12.17
N_{km}	0.11	0.58	0.37
Flexion Moment (Nm)	2.52	27.44	11.08
Extension Moment (Nm)	0.50	11.8	7.68
Tension (N)	1	500	151
Compression (N)	39	372	149

RETRACTION PHASE SENSITIVITY: In the retraction phase, NIC was found to be most sensitive to crash duration, which in this study defined the severity of the crash pulse in terms of duration/peak acceleration. Other factors that were not modeled could also be expected to influence NIC, including seat geometry and friction between the occupant and the seat. NIC also showed significant sensitivity to head restraint backset. Seat stiffness and head restraint height had little influence on NIC. The influence of all four design factors on the retraction phase, as represented by the NIC criterion, is shown in Figure 2.

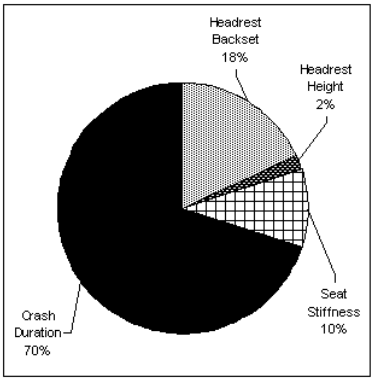


Figure 2: Retraction Phase Sensitivity

EXTENSION PHASE SENSITIVITY: In most of the runs, a true extension phase did not occur. Neck extension moment was used to assess the injury potential for this phase. Head restraint backset had the greatest influence on neck extension moment. Extension displayed some sensitivity to crash duration, but very little to head restraint height or seat stiffness. The influence of all four design factors on the extension phase, as represented by extension moment, is shown in Figure 3.

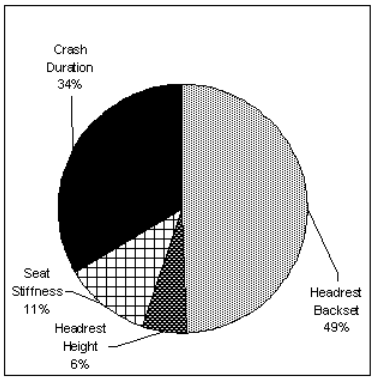
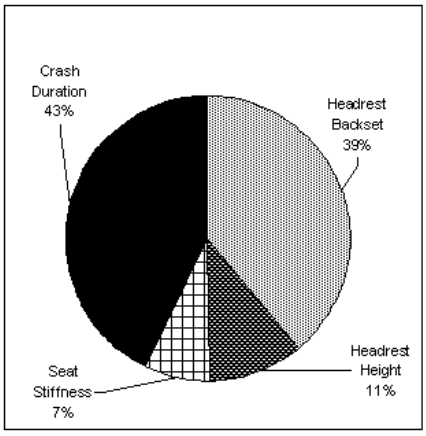


Figure 3: Extension Phase Sensitivity

REBOUND PHASE SENSITIVITY: In the rebound phase, crash duration showed the most influence on N_{km} followed closely by head restraint backset. Head restraint height and seat stiffness had minimal influence on N_{km}, although seat stiffness had a significant influence on one of the four measures making up the N_{km} criterion. For N_{km} Flexion Posterior, seat stiffness had an influence of 18%, an influence that was lost when combined into a maximum N_{km} value. Also, seat stiffness had an influence of 17 % on the shear force posterior measure.



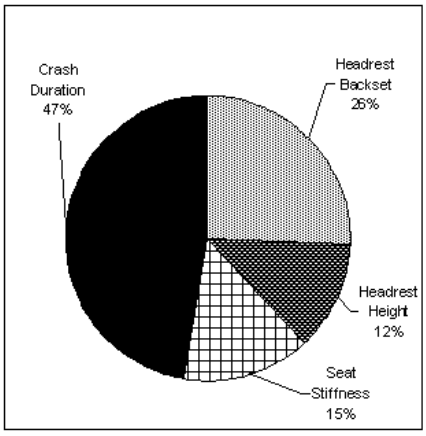


Figure 4: Rebound Phase Sensitivity (Left) and Protraction Phase Sensitivity (Right)

PROTRACTION PHASE SENSIVITIY: In the protraction phase, measured by flexion moment, crash duration emerged as the most significant influence on injury potential. Head restraint backset had an influence of 26 %, while seat stiffness and head restraint height had an influence of 15 % and 12 %, respectively.

ANALYSIS AND DISCUSSION

These results suggest that vehicle design plays an important role in protecting the occupant from whiplash injury and highlights the importance of examining the effect of all key design features on the different phases of the whiplash motion. Different design factors may have different levels of influence on the phases of the whiplash motion.

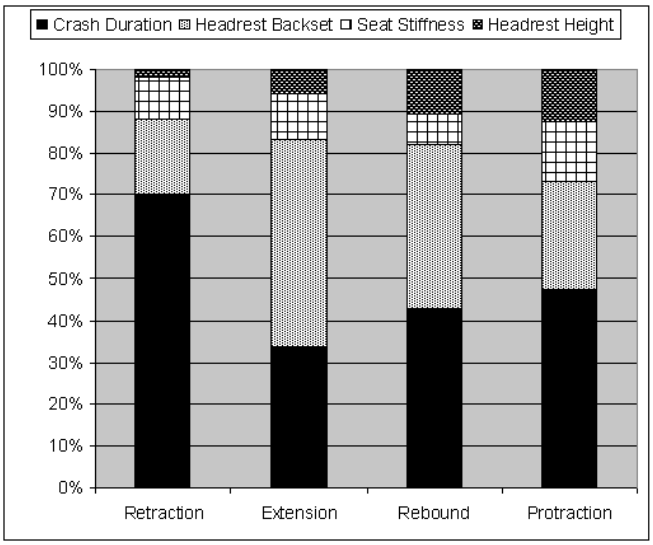


Figure 5: Design Sensitivity by Whiplash Phase

The results also suggest that, in designing and rating vehicles for whiplash protection, it is important to look at the vehicle as a system and not just a single design variable such as seat stiffness or head restraint geometry. It is likely that, at different crash magnitudes, the influence of vehicle design parameters would change significantly. For example, seat stiffness showed to have little influence on the injury measures for any phase of motion (although it did have a significant influence on one of the four phases of N_{km}). The influence of seat properties at higher impact magnitudes is well established, and it is likely that the relatively low magnitude of this study was simply not enough to significantly invoke the seat stiffness.

The results of this study also suggest that the multiple phases of whiplash make vehicle design quite challenging. Even with this greatly simplified model, the relationship between design features and proposed injury measures is shown to be quite complex. It is likely that examination of other parameters known to influence occupant motion would only add to the complexity demonstrated by this relatively simple study.

The results also highlight the influence on structural force-deflection characteristics in mitigating whiplash. The significant role that vehicle stiffness plays, particularly in the retraction phase, is something that should be considered by testing agencies that will be rating vehicle performance.

Key conclusions and recommendations for rear impact testing designed to measure whiplash protection:

- Given the strong influence of bumper design on whiplash injury potential, test agencies should consider incorporating vehicle-specific crash pulses into their sled testing.
- Whiplash tests should ideally be performed at more than one crash magnitude to get a more precise understanding of design performance.
- Computer simulation should be considered as a supplement to physical crash testing for assessing whiplash performance.

LIMITATIONS AND FURTHER WORK

This study was based on a very simple impact scenario, a well-aligned rear impact with no underride, override, angle or offset. Other factors are also known to influence that motion that were not varied, including seat angle [16], occupant/seat friction [34], seat cushion stiffness [16, 48], head

restraint stiffness [16, 48] and occupant characteristics [20], position [34, 27, 28] and occupant reaction [4, 26, 47]. It is likely that consideration of these additional factors would have resulted in greater variation. This paper was also limited to an impact magnitude of 10 kph. It is hypothesized that the influence of the design factors on the different phases of occupant motion will vary significantly based on impact magnitude. Comparison of the influence of key design factors on the various phases of whiplash at varying impact magnitudes is reserved for a future work. Additionally, this paper did not sort out the affects that are purely within the control of the vehicle manufacturer from those that are in control of the occupant, also reserved for a future study.

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APPENDIX A: OCCUPANT MODEL RESULTS COMPARISON

To establish a level of confidence on the kinematics of the occupant model, we compared the results obtained from the model with a set of 43 rear-impact sled tests conducted using 19 human volunteers [5]. The seat backrest was oriented at 25 degrees backward inclination and a regular 3-point belt was used. The performance of the model on various measures is presented in Figure 6 to Figure 8. The lower and upper corridors from live human tests are plotted along with the average human response. Hybrid III and RID2 responses are also shown on each plot.

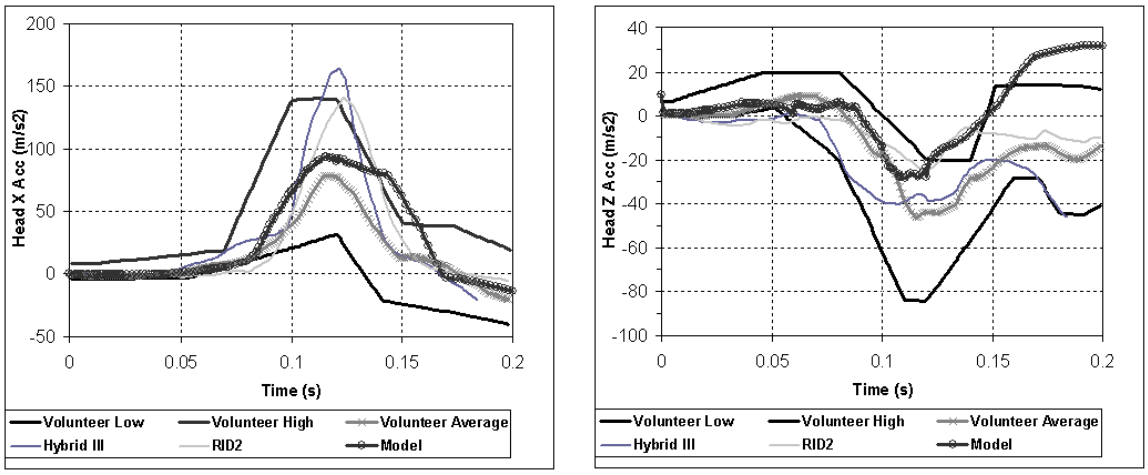

Figure 6: Validation Results on Head X Acceleration (left) and Head Z Acceleration (right)

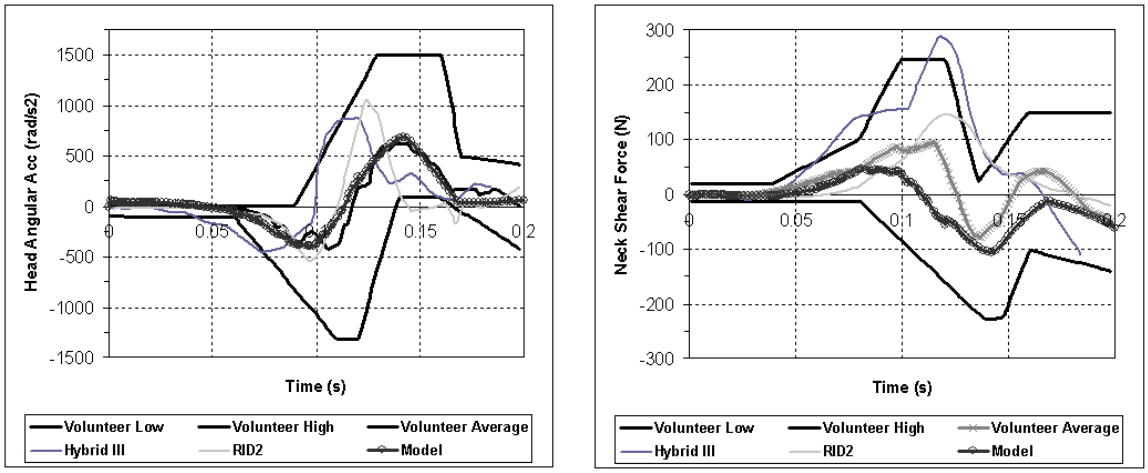

Figure 7: Validation Results on Head Angular Acceleration (left) and Neck Shear Force (right)

Model response fits reasonably well within the lower and upper corridors, especially the peak point of these outputs, which are considered important on injury potential [3, 37, 30, 36]. Also, the "shape" of the outputs is "similar" to the occupant responses from the measurements. This shows that model has bio-fidelity similar to live human subjects, some of which the Hybrid III and RID2 ATDs fail to capture. Differences between the model and data can be attributed to the uncertainty in the parameters used in the tests, particularly the head restraint position, and variations in neck strength and muscle response times [4, 26, 47] of the various volunteers.

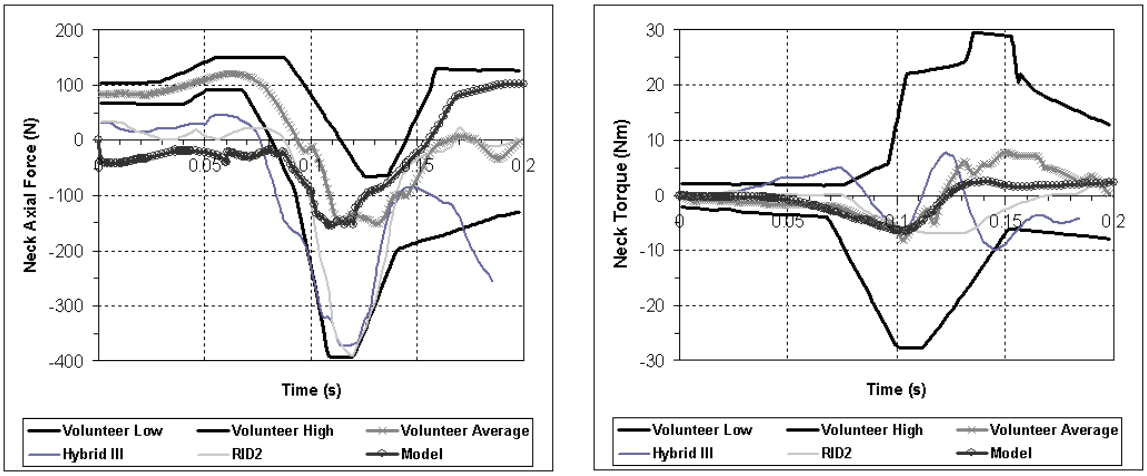

Figure 8: Validation Results on Neck Axial Force (left) and Neck Torque (right)

APPENDIX B: IDEALIZED CRASHPULSES

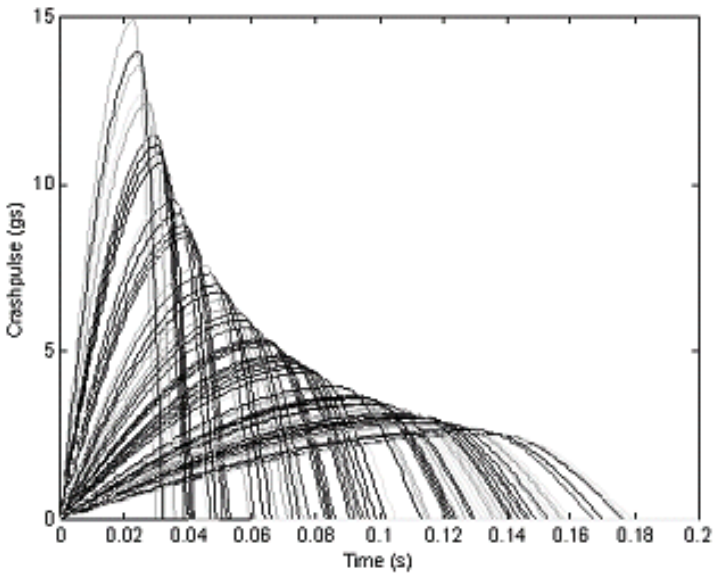

Figure 9: Idealized Crashpulses Used in Monte Carlo Simulations

APPENDIX C: ADDITIONAL FIGURES ON MONTE CARLO ANALYSIS

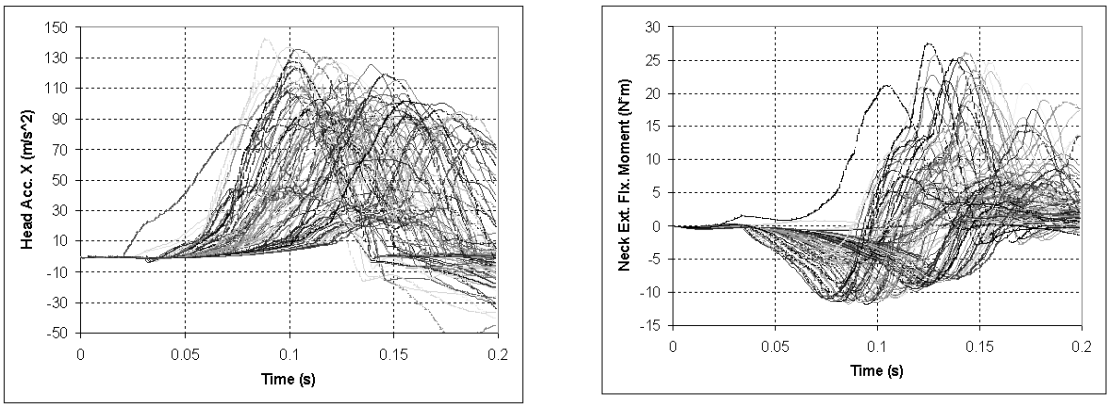

Figure 10: Time Histories for Head Acceleration X (left) and Extension/Flexion Moment (right)

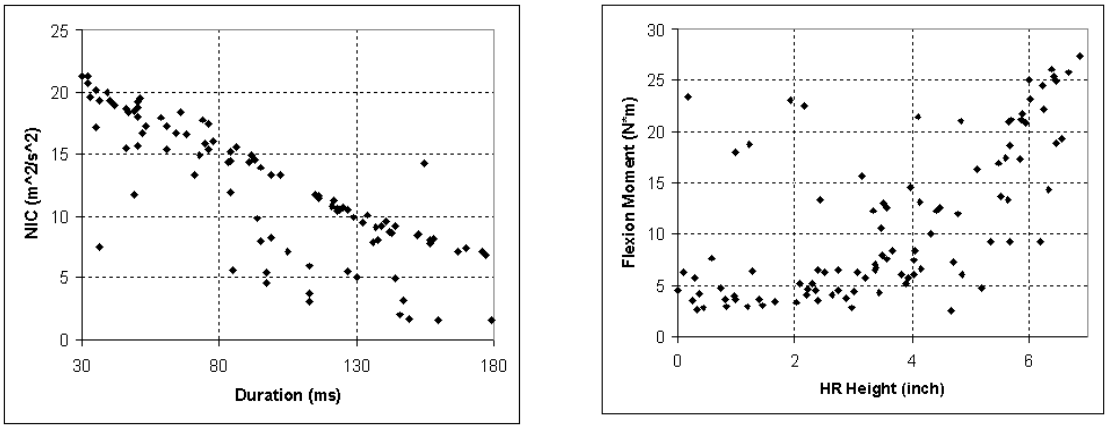

Figure 11: NIC versus Duration (left) and Flexion Moment versus Headrest Height (right)

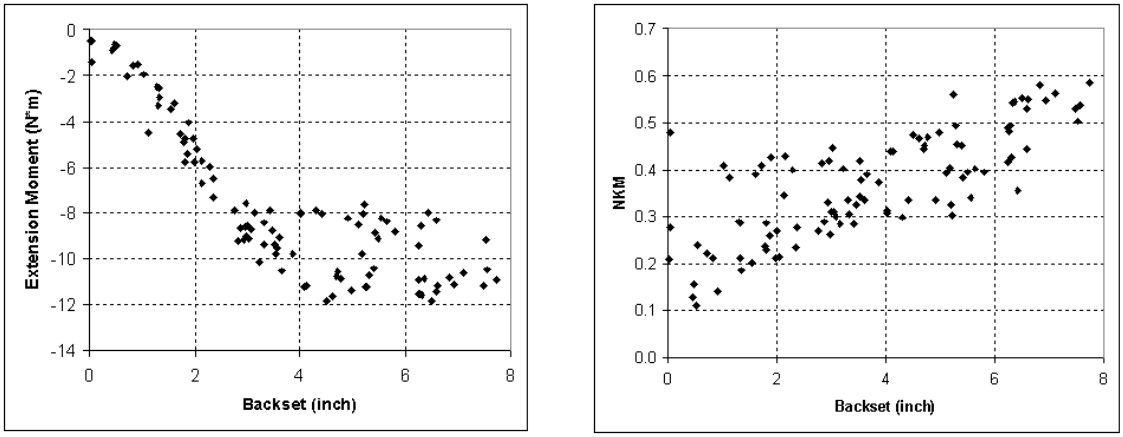

Figure 12: Extension Moment versus Backset (left) and Nkm versus Backset (right)