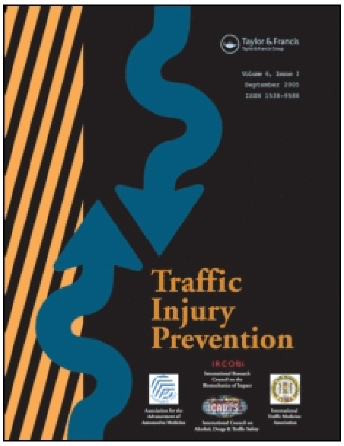


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# Incorporation of Lower Neck Shear Forces to Predict Facet Joint Injury Risk in Low-Speed Automotive Rear Impacts

Brian D. Stemper[abc]; Steven G. Storvik[abc]
[a] Department of Neurosurgery, Medical College of Wisconsin, Milwaukee, Wisconsin [b] Department of Biomedical Engineering, Marquette University, Milwaukee, Wisconsin [c] Veterans Affairs Medical Center, Milwaukee, Wisconsin

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# Incorporation of Lower Neck Shear Forces to Predict Facet Joint Injury Risk in Low-Speed Automotive Rear Impacts

**BRIAN D. STEMPER**[1,2,3] **and STEVEN G. STORVIK**[1,2,3]

[1]Department of Neurosurgery, Medical College of Wisconsin, Milwaukee, Wisconsin
[2]Department of Biomedical Engineering, Marquette University, Milwaukee, Wisconsin
[3]Veterans Affairs Medical Center, Milwaukee, Wisconsin



*Lower neck shear force remains a viable candidate for a low-velocity automotive rear-impact injury criterion. Data were previously reported to demonstrate high correlations between the magnitude of lower neck shear force and lower cervical spine facet joint motions. The present study determined the ability of lower neck shear force to predict soft-tissue injury risk in simulated automotive rear impacts. Rear-impact tests were conducted at two velocities and with two seatback orientations using a Hybrid III anthropomorphic test device (ATD) and stock automobile seats from 2007 model year vehicles. Higher velocities and more vertical seatback orientations were associated with higher injury risk based on computational modeling simulations performed in this study. Six cervical spine injury criteria including NIC, Nij, Nkm, LNL, and lower neck shear force and bending moment, increased with impact velocity. NIC, Nij, and shear force were most sensitive to changes in impact velocity. Four metrics, including Nkm, LNL, and lower neck shear force and bending moment, increased for tests with more vertical seatback orientations. Shear force was most sensitive to changes in seatback orientation. Peak values for shear force, NIC, and Nij occurred approximately at the time of head restraint contact for all four test conditions. Therefore, of the six investigated metrics, lower neck shear force was the only metric to demonstrate consistency with regard to injury risk and timing of peak magnitudes. These results demonstrate the ability of lower neck shear force to predict injury risk during low velocity automotive rear impacts and warrant continued investigation into the sensitivity and applicability of this metric for other rear-impact conditions.*

**Keywords** Whiplash; Rear impact; Crash dummies; Seats, Injury mechanism; Biomechanics

## *INTRODUCTION*

Correlation of lower neck shear force with facet joint shear and distraction motions was investigated with the goal of developing a lower neck low-speed rear-impact injury criterion (Stemper et al. 2007). The analysis demonstrated that lower cervical facet joint resultant motions were correlated to shear forces measured at the cervico-thoracic junction in a series of simulated rear impacts using postmortem human subject (PMHS) head–neck complexes, indicating that lower neck shear force can be used to predict cervical spine facet joint ligamentous distortion during low-velocity rear impacts. The finding was significant due to the previous biomechanical implication of lower cervical facet joints in the low-speed rear-impact injury mechanism (Cusick et al. 2001; Dehner et al. 2007; Deng et al. 2000; Stemper et al. 2004a; Sundararajan et al. 2004; Winkelstein et al. 2000; Yoganandan et al. 2001) and the fact that ligaments fail due to tension (Myklebust et al. 1988). However, lower neck shear force remains unproven as an injury criterion. To be adopted into existing automobile safety assessment protocols such as the newly enacted EuroNCAP Dynamic Assessment of Car Seats for Neck Injury Protection (EuroNCAP; European New Car Assessment Programme 2009) or the United States Department of Transportation (DOT) Federal Motor Vehicle Safety Standard No. 202 (FMVSS No. 202; National Highway Traffic Safety Administration 2004), the metric must be proven to reliably predict soft-tissue cervical spine injuries and demonstrate sensitivity to impact and occupant related factors that affect injury risk.

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Address correspondence to: Brian D. Stemper, PhD, Associate Professor, Department of Neurosurgery, Medical College of Wisconsin, 5000 West National Ave., Research 151, Milwaukee, WI 53295. E-mail: bstemper@mcw.edu

Although a controlled PMHS study has not been conducted with the intention of correlating lower neck shear forces to soft tissue cervical spine injuries during low-velocity rear impacts, data exist in literature that can be used for this purpose. A limited number of studies have reported lower neck shear forces during experimental or computational simulation of automotive rear impacts (Table I). Some of those studies have commented

**Table I** Rear impact studies reporting lower neck shear forces

| First author(s) | Year(s) | Test subjects | Head restraint | Impact | Lower neck shear (N) |
|---|---|---|---|---|---|
| Jakobsson | 1994 | MADYMO | Model | 11 km/h | 75 |
| Benson | 1996 | Hybrid III | Stock | 34 km/h | 654–2148 |
| Prasad | 1997 | Hybrid III, RID neck | N/A, Stock | 8, 16, 24 km/h | 275–700 |
| Yoganandan, Philippens | 2000–2001 | PMHS | N/A | 15–25 km/h | 331–809 |
| Tencer | 2002 | MADYMO | Model | 12 *g* | 150 |
| Kim | 2003 | Hybrid III, RID2 | N/A | 17, 28 km/h | 300–638 |
| Stemper | 2003 | Head-neck PMHS | N/A | 2–9 km/h | 84–246 |
| Cappon | 2005 | RID3D, BioRID II | Stock | 16 km/h | 75–460 |
| Latchford | 2005 | BioRID II | Stock | 10 km/h | 265–348 |

on lower neck shear force with regard to injury risk in specific rear impact orientations. The studies have generally reported positive results with regard to the ability of lower neck shear force to predict injurious conditions.

Two studies were performed to investigate the effect of occupant seating position on neck loads and injury criteria. The first study employed the Hybrid III anthropomorphic test device (ATD) and stock automobile seats tested with the ATD in normal and pitched forward positions (Benson et al. 1996). The second study tested the BioRID ATD in a single stock automobile seat with seatback angles between 20 and 30 degrees from vertical (Latchford et al. 2005). These two protocols resulted in opposite trends with regard to lower neck shear forces. Benson et al. (1996) reported up to a 196 percent increase in lower neck shear forces with occupants pitched forward, whereas Latchford et al. (2005) reported decreases in lower neck shear forces and all injury criteria with more vertically oriented seats. Though both studies demonstrated lower neck shear force sensitivity to changes in occupant position during rear impacts, results with regard to the ability of the metric to predict neck injuries were not conclusive.

A third study correlated real-world accident data with seat specific rear impact performance as measured during horizontal sled tests (Cappon et al. 2005). Insurance claims were used to derive vehicle-specific data on occupant protection in single rear impacts. Those data were correlated to results of rear-impact sled tests using the RID3D and BioRID II ATDs conducted at 16 km/h using standard and vehicle-specific acceleration versus time pulse shapes. Seat performance was assessed using cervical spine injury criteria including NIC, Nkm, LNL, Nij, and upper and lower neck loads. Real-world accident data correlated best with lower neck shear force for the RID3D ATD, with $R^2$ values of 0.79 and 0.80 for the standard and vehicle-specific pulses. This led the authors to indicate that lower neck shear force was a potential candidate for a relevant injury criterion. Correlation for lower neck shear, as well as all other criteria, was lower in the BioRID II ATD. However, the authors acknowledged that in this earlier version of the ATD, the lower neck load cell was only recently incorporated and that biofidelity of lower neck readings had not been confirmed. However, results of that analysis confirmed that lower neck shear force is a viable candidate for a low-speed rear-impact injury criterion.

Most relevant to the present investigation, Yoganandan, Philippens, and colleagues subjected one male and four female PMHS to single exposure rear impacts using a rigid seat (Philippens et al. 2000; Yoganandan et al. 2000, 2001). Rear-impact velocities were approximately 15 or 25 km/h (Table II). One of two PMHS subjected to low-velocity rear impacts and all three PMHS subjected to high-velocity rear impacts sustained posterior spinal column injuries. Three of four specimens sustained facet joint injury at C4-C5, C5-C6, or C6-C7. According to our previous investigation (Stemper et al. 2007), the lower neck shear force threshold for facet joint ligamentous injury was 636 N in males and 384 N in females. Injuries in three of four PMHS sustaining injury would have been predicted by the lower neck shear force limit (H1, H2, and H3). Additionally, lower neck shear force in the specimen not sustaining posterior column injury (L1) was below the threshold for females. Therefore, in this limited test series, lower neck shear force proved to be a valid predictor of posterior column soft-tissue cervical spine injuries sustained during simulated rear impacts.

The purpose of the present study was to determine the ability of lower neck shear force to predict soft-tissue injury risk during low-speed automotive rear impacts. A unique methodology was employed to measure lower neck shear forces in the Hybrid III ATD and quantify facet joint ligament strains in a comprehensively validated computational model during simulated rear impacts with expected outcomes. In particular, tests were

**Table II** PMHS test series reporting lower neck loads and soft-tissue spinal injuries

| PMHS | Gender | Delta-V (km/h) | Lower neck shear force (N) | Posterior column injuries |
|---|---|---|---|---|
| L1 | Female | 14.9 | 331 | None |
| L2 | Female | 15.9 | 372 | C5-6 Facet joint widening |
| | | | | C6-7 Ligamentum flavum tear |
| H1 | Female | 24.8 | 571 | C6-7 Ligamentum flavum rupture |
| H2 | Female | 24.5 | 675 | C5-6 Facet joint hematoma |
| H3 | Male | 23.7 | 809 | C4-5 Facet joint widening |
| | | | | C5-6 Facet joint disruption |
| | | | | C5-6 Facet joint capsule stretch |

conducted at two impact velocities and with two seatback angles. Increasing impact velocity has been associated with greater injury risk in rear impact (Ono et al. 1997; Stemper et al. 2002; Szabo and Welcher 1996). Accordingly, lower neck shear force and facet joint ligament strains should increase at higher impact velocities. Seatback orientation may also influence injury risk in rear impacts. For a given head restraint backset, more vertically oriented seatbacks should increase the magnitude of lower neck shear force, according to vector mechanics. For shear force to be a valid predictor of injury, facet joint ligament strains should also increase with seatback verticality. Literature has supported this assertion by demonstrating greater spinal motions, resulting in increased soft-tissue strains, during tests with occupants in more vertical seated positions (Deng et al. 2000; Pramudita et al. 2007; Sundararajan et al. 2004). The ability of lower neck shear force to predict greater facet joint ligament strains was compared to lower neck bending moment and accepted injury criteria including the NIC, Nkm, Nij, and LNL.



## *METHODS*

The study methodology was composed of two parts. The first part consisted of experimentally simulating automotive rear impacts using stock automobile seats and a Hybrid III 50th percentile ATD. Four tests were conducted to investigate the effects of impact velocity and seatback orientation on the magnitude and timing of existing injury criteria and lower neck loads. These metrics were compared based on the expected outcomes that injury risk increases with greater impact velocities and seatback orientations closer to vertical. The second part of the study consisted of quantifying facet joint ligament strains using a comprehensively validated head–neck computational model to verify assumed injury risks from the experimental tests. Rear impact was simulated in the model by accelerating the first thoracic vertebra (T1) using horizontal and vertical accelerations measured in the Hybrid III ATD during the first part of the study. The analysis compared the magnitude and timing of existing injury criteria and neck loads obtained from experimental testing with relative injury risks to determine the sensitivity of these metrics to changes in impact conditions.

### *Experimental Testing*

Four rear-impact tests were conducted using a 50th percentile male Hybrid III ATD, stock automobile seats, and a ServoSled catapult sled system (Seattle Safety, Kent, Wash). The test matrix was designed to quantify sensitivity of existing injury criteria and lower neck shear forces to rear impact velocity and seatback orientation. Two stock automobile seats obtained from identical top-selling 2007 model year vehicles were mounted to the sled. Seatback angle was adjusted to 25 or 15 degrees from vertical. A three-point seat belt was used to prevent excessive forward rebound from the seatback following rear impact. Consistency in ATD seating position was obtained using a 50th percentile H-point machine. Head restraint backset was measured using an ICBC head restraint measuring device and confirmed once the ATD was positioned in the seat. Instrumentation included upper neck and lower neck six axis load cells, tri-axial accelerometer arrays at the head center of mass and upper thorax, and redundant accelerometers on the sled base. Biomechanical data were anti-alias filtered, recorded at 12.5 kHz, and low-pass filtered according to SAE specifications (Society of Automotive Engineers [SAE] 1995).

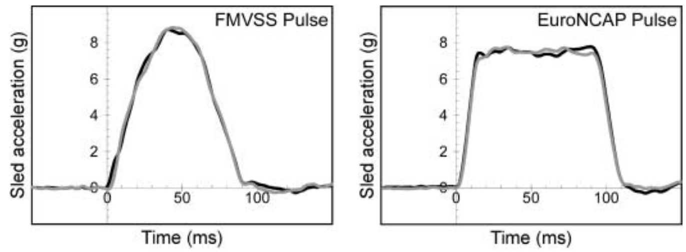


**Figure 1** Acceleration versus time for lower (FMVSS) and higher (EuroNCAP) velocities (25 degree seatbacks: darker lines, 15 degree seatbacks: lighter lines).

The test matrix consisted of lower and higher velocity tests. Acceleration versus time pulse shape for the lower velocity test was designed to match the dynamic head restraint assessment pulse outlined in FMVSS No. 202 (NHTSA 2004); the shape for the higher velocity test was designed to match the high velocity EuroNCAP whiplash pulse (European New Car Assessment Programme 2009; Figure 1). One test was conducted at each acceleration pulse with seatbacks oriented at 15 and 25 degrees, for a total of four tests. Each seat was subjected to one lower velocity and one higher velocity test.

During each test, the following biomechanical signals were recorded and used to compute rear-impact injury metrics: horizontal acceleration of the head, horizontal and vertical accelerations of the upper thorax (T1), upper neck anterior-posterior shear force, upper neck tension-compression force, upper neck sagittal plane bending moment, lower neck anterior-posterior shear force, lower neck tension-compression force, lower neck sagittal plane bending moment, and lower neck coronal plane bending moment. Time of head restraint contact was measured using videographic data obtained from the lateral side of the ATD. The following injury metrics were computed and used to compare between impact velocities and seatback orientations: NIC (Bostrom et al. 1996), Nkm (Schmitt et al. 2002), Nij (Nte; Klinich et al. 1996; Mertz and Prasad 2000; Prasad and Daniel 1984), LNL (Heitplatz et al. 2003), maximum lower neck anterior-posterior shear force (Fx), and maximum lower neck bending moment (My). NIC values were computed during the first 150 ms following initiation of acceleration. Maximum values and times of occurrence for existing injury criteria, lower neck shear force, and sagittal plane bending moment were compared between the four tests to determine the effects of impact velocity and seatback orientation on predicted injury risk.

Repeatability of stock automobile seats was assessed in a separate set of experiments. A single seat not used in the testing described above was subjected to four tests of approximately equal changes in velocity. Lower neck Fx and MY and NIC, Nkm, Nij, and LNL were obtained following each test. Peak values were compared between the four tests and the coefficient of variation was computed as the ratio of standard deviation to mean to quantify the variability between tests.

### *Computational Modeling*

Experimental sled tests were simulated using a previously validated head–neck MADYMO model (Stemper et al. 2004b). The model consisted of a rigid head, seven cervical vertebrae, and first thoracic vertebra. Cervical spine soft tissues were modeled using discrete elements with nonlinear and viscoelastic material properties obtained from literature. Cervical spinal ligaments were modeled using tension-only Kelvin restraints, with level dependent response. Four Kelvin restraints were used to model each facet joint capsular ligament and placed in anterior, lateral, posterior, and medial joint regions. Bending responses of intervertebral joints were modeled using level specific moment-rotation responses. Intervertebral and facet joint shear and tension/compression responses were modeled using level specific force-displacement responses. Sixteen neck muscle pairs were modeled using 136 Hill-type elements, with passive and active properties. Validation consisted of comparing overall motions (i.e., head to T1), segmental angulations, and lower cervical facet joint motions at multiple rear-impact velocities to corridors obtained from PMHS.

Initial position of the model was controlled specifically to represent a normal subject during an unexpected rear impact. Because cervical posture was shown to affect facet joint ligament distractions during rear impact (Stemper, Yoganandan, and Pintar 2005), lordotic curvature of the model was designed to match the mean lordosis of 48 normal volunteers (Takeshima et al. 2002). The Frankfort plane of the skull was oriented horizontally and the T1 vertebra was given an anterior orientation of 25 degrees. A head restraint with 5 cm backset was added to resist head–neck hyperextension. Vertical position of the head restraint was adjusted such that the head would contact the center of the restraint in all simulations. Posterior-anterior (x-axis) and inferior-superior (z-axis) acceleration versus time pulses measured at the upper thorax (i.e., T1) in the Hybrid III ATD during experimental testing were used to drive the computational model. Lower velocity T1 x- and z-axis acceleration pulses for 25 and 15 degree seatback orientations are presented in Figure 2. Higher velocity T1 x- and z-axis acceleration pulses for 25 and 15 degree seatback orientations are presented in Figure 3. The model was exercised once for each experimental pulse. Sagittal plane rotation of the T1 vertebra was controlled using a rotational stiffness coefficient derived previously (Stemper et al. 2005a). Active neck muscle contraction was simulated using a reflex contraction paradigm published previously and incorporating 50 ms reflex delay, 13 ms electromechanical delay, and 81 ms muscle force rise time (Stemper et al., 2005b).

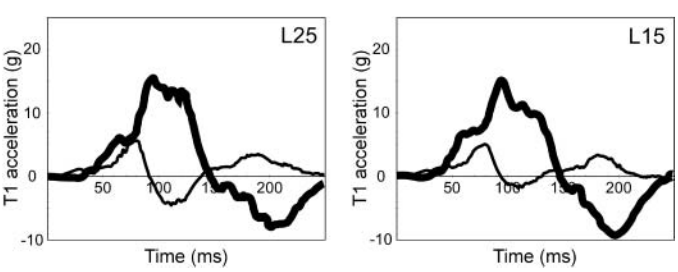


**Figure 2** Lower velocity T1 acceleration pulses (x-axis: dark lines, z-axis: light lines).

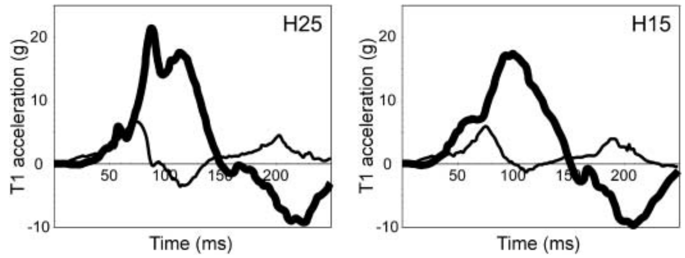


**Figure 3** Higher velocity T1 acceleration pulses (x-axis: dark lines, z-axis: light lines).

Facet joint ligament strains (mm/mm) were quantified on right and left sides, in each facet joint region, and at each spinal level from C2-C3 through C6-C7 from the initiation of T1 acceleration until the head rebounded from the head restraint. Strain was computed as the change in length of the element divided by the initial length. Maximum strains were recorded at each level and were used to demonstrate varying levels of injury risk following the four experimental tests, under the assumption that increasing strain results in greater injury potential. Maximum strains were compared to injury metrics computed following experimental tests to determine which injury metrics correlated best with risk of soft-tissue cervical spine injury.

## *RESULTS*

### *Experimental Testing*

Four tests were conducted using two automobile seats. Lower velocity tests were conducted at 17.6 km/h change in velocity (Table III). Higher velocity tests were conducted at 24.8 km/h change in velocity. Sled acceleration versus time characteristics fit within the suggested limits for FMVSS No. 202 and EuroNCAP. Head restraint backset was approximately the same between 25 and 15 degree seatback orientations due to the inflexible shape of the Hybrid III torso. Head-to-head restraint contact occurred between 81 and 84 ms for all four experimental tests. Lower neck bending moments and shear forces were measured during and injury criteria were computed following each experimental test. Peak magnitudes and the associated time are presented in Table IV.

Sensitivity of each metric to changes in rear impact conditions can be determined by comparing peak magnitudes from each test to the experimental test with lowest projected injury risk. According to the assumptions of this study, the lowest injury risk would result from the lower velocity test with the 25 degree seatback (i.e., L25). Peak magnitudes of injury criteria and lower neck loads from the other three tests were compared to the baseline test (Figure 4). Magnitude of all metrics increased with greater rear-impact velocity for 25 and 15 degree seatback orientations. Greatest average sensitivity to rear-impact velocity for both seatback orientations was demonstrated by NIC (38%), Nij

**Table III** Seat and pulse characteristics for experimental testing

| Test | Seatback angle (degree) | Delta-V (km/h) | Mean acceleration ($m/s^2$) | Maximum acceleration ($m/s^2$) | Head restraint backset (cm) |
|---|---|---|---|---|---|
| L25 | 25 | 17.6 | 44.1 | 85.3 | 5.0 |
| H25 | 25 | 24.9 | 62.8 | 76.3 | 6.0 |
| L15 | 15 | 17.5 | 49.3 | 86.5 | 5.0 |
| H15 | 15 | 24.7 | 62.3 | 76.1 | 5.0 |

(33%), and Fx (27%). With regard to seatback orientation, peak magnitude of Nkm, LNL, Fx, and My increased with 15 degree seatback orientations for both impact velocities. Peak magnitude of NIC and Nij increased with the 15 degree seatback for lower velocity tests but decreased with the 15 degree seatback orientation for the higher velocity tests. Greatest average sensitivity to seatback orientation for both velocities was demonstrated by Fx (42%), LNL (23%), and Nkm (17%). Timing of maximum values also varied between metrics. Peak Fx, NIC, and Nij occurred at approximately the time of head restraint contact for all four tests. Peak My occurred during the rebound phase for all four tests. Trends were inconsistent with regard to the timing of peak Nkm and LNL, occurring early in some cases and late in others.

Four separate tests were conducted to determine the repeatability of the Hybrid III and stock automobile seats. Mean sled change in velocity was 9.7 ± 0.3 km/h. Mean values for Fx, My, NIC, Nkm, Nij, and LNL were 337.3 ± 23.6 N, 52.1 ± 2.6 Nm, 9.3 ± 0.4 $m^2/s^2$, 0.39 ± 0.02, 0.08 ± 0.01, and 4.3 ± 0.3, respectively. All six metrics demonstrated excellent repeatability across the four tests, with coefficients of variation of 7.0 percent or less for five of the six metrics. Coefficient of variation for Nij was 15.2 percent.



### Computational Modeling

The head–neck computational model was exercised using T1 acceleration pulses measured in the Hybrid III during the first part of the study. The model demonstrated the characteristic phases of retraction, extension, and rebound. However, response of the model beyond the time of head restraint contact has not been validated and data are only presented from the initiation of T1 acceleration until shortly after head restraint contact. Facet joint ligament strains increased until the head contacted the head restraint or shortly after (Figure 5). Strains were greatest at lower cervical levels and decreased cranially to C3-C4.

Peak facet joint ligament strains demonstrated dependence upon impact velocity and seatback orientation (Table V). Regions of the facet joint ligament that sustained the highest levels of strain were the posterior region at the C2-C3 level and lateral or medial regions for all caudal levels (C3-C4 through C6-C7). Peak ligament strains increased with impact velocity for both seatback orientations and at all spinal levels, except C4-C5 wherein strains were equal for lower and higher velocity tests with 25 degree seatback. Ligament strains were greater at 15 compared to 25 degree seatback angles at all levels and velocities except the C4-C5 level during the lower velocity simulation. Ligament strains increased an average of 20.0 percent (−2.1 to 71.4%) for 15 degree seatback orientations. These findings confirm the earlier assumptions of increased injury risk for higher velocity impacts and more vertical seatback orientations.

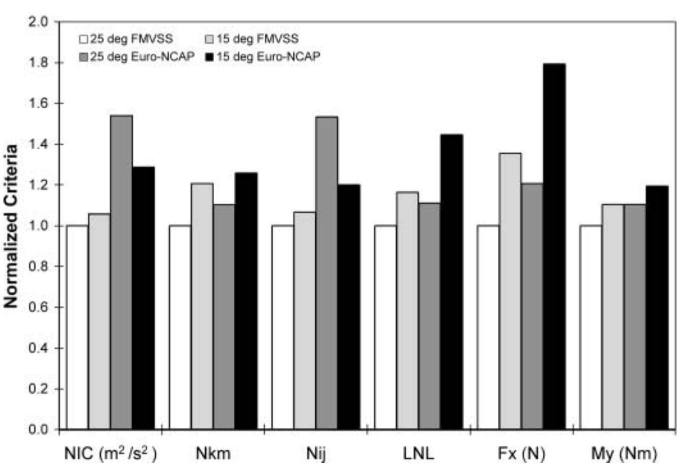


**Figure 4** Sensitivity of injury criteria and lower neck loads to changes in rear impact conditions. All criteria are normalized relative to the baseline value obtained in the lower velocity test with 25 degree seatback.

## DISCUSSION

Results of this study demonstrated the viability of lower neck shear force as a possible low-velocity rear-impact injury criterion. Experimental tests were conducted at two impact velocities and with two seatback orientations. Data from literature were used to determine relative injury risk between the different test conditions, with higher velocities and more vertical seatback orientations associated with higher injury risk. Present computational modeling results confirmed these assumptions by demonstrating greater facet joint ligament strains for test conditions assumed to have higher injury risk. Four of the six investigated metrics demonstrated consistency with regard to injury

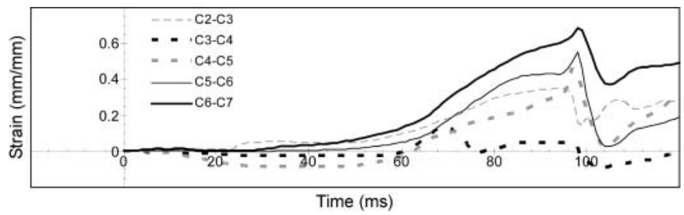


**Figure 5** Representative plot of level-by-level facet joint ligament strains versus time.

**Table IV** Injury metrics and lower neck kinetics obtained during experimental testing

| Test | NIC ($m^2/s^2$) | Nkm | Nij | LNL | Fx (N) | My (Nm) |
|---|---|---|---|---|---|---|
| L25 | 13.9 (130) | 0.58 (248) | 0.15 (104) | 5.43 (250) | 513 (104) | 67 (251) |
| H25 | 21.4 (81) | 0.64 (262) | 0.23 (100) | 6.03 (101) | 619 (102) | 74 (264) |
| L15 | 14.7 (82) | 0.70 (107) | 0.16 (106) | 6.32 (102) | 695 (104) | 71 (238) |
| H15 | 17.9 (78) | 0.73 (107) | 0.18 (107) | 7.85 (107) | 920 (106) | 80 (245) |

*Numbers in parentheses represent the time of occurrence in milliseconds.



risk: Nkm, LNL, lower neck shear force, and lower neck bending moment. Shear force demonstrated the highest sensitivity of the four to changes in impact velocity and seatback orientation, increasing by an average of 27 percent for a 7.2 km/h increase in velocity and by 42 percent for a 10 degree change in seatback orientation. Likewise, the six metrics were also evaluated based on the timing of peak magnitudes. Soft-tissue injuries are generally assumed to occur prior to or at the time of head restraint contact during low-velocity automotive rear impacts. This assumption can be confirmed by an abundance of evidence indicating that injury risk increases with head restraint backset (Linder et al. 2001; Stemper et al. 2006a; Szabo et al. 2003; Tencer et al. 2001). With regard to timing, only NIC, Nij, and shear force consistently attained peak values prior to or shortly following head restraint contact. Therefore, lower neck shear force was the only metric to demonstrate consistency with regard to injury risk and timing of peak magnitudes. Although the use of lower neck shear force as an injury criterion was verified in the present study, results are by no means comprehensive. Continued validation of the metric with regard to injury risk is required. However, results of this study warrant the quantification of lower neck shear force as an injury predictor in future experimental testing and computational modeling.

The focus of this study was on the influence of impact velocity and seatback orientation on the magnitude of lower neck shear force during simulated automotive rear impacts. Increasing impact velocity (i.e., delta-V) was expected to increase the magnitude of lower neck shear force. Likewise, based on vector mechanics, a more vertically oriented seatback was also expected to increase lower neck shear force. Those two factors were chosen as the focus of the present study due to their discernable effect. However, other factors are likely to influence shear force and injury potential during automotive rear impacts. Because the magnitude of lower neck shear force is primarily modulated by the retraction phase, characteristics of the head restraint (e.g., backset, shape, and material properties) will likely also have a considerable influence on this metric. Occupant-related factors, such as body size and vertebral alignment at the time of impact, will also likely influence shear force. However, a controlled biomechanical investigation of these factors is required to delineate their effects.

**Table V** Maximum facet joint ligament strains (mm/mm)

| | C2-C3 | C3-C4 | C4-C5 | C5-C6 | C6-C7 |
|---|---|---|---|---|---|
| L25 | 0.35 (94) | 0.14 (69) | 0.47 (97) | 0.55 (98) | 0.69 (98) |
| H25 | 0.39 (94) | 0.23 (93) | 0.47 (95) | 0.63 (95) | 0.80 (96) |
| L15 | 0.37 (91) | 0.24 (71) | 0.46 (94) | 0.62 (94) | 0.75 (95) |
| H15 | 0.41 (86) | 0.37 (70) | 0.52 (68) | 0.77 (89) | 0.84 (90) |

Numbers in parentheses indicate timing (ms) of the maximum value.

Although lower neck shear force proved to be sensitive to changes in seatback orientation (Latchford et al. 2005) and occupant orientation relative to the seatback (Benson et al. 1996; Sundararajan et al. 2004) in the present and previous investigations, trends were somewhat contradictory. Present results are supported by Benson et al. (1996) in that higher shear forces resulted when occupants were seated more vertically relative to the seat, whereas Latchford et al. (2005) reported decreased shear forces for more vertically oriented seatbacks. Increased facet joint ligament strains for more vertical seating positions obtained during the computational part of the present study are supported by facet joint stretch magnitudes in PMHS reported by Sundararajan et al. (2004) and indicate increased injury risk for more vertical seating positions. A possible explanation for differences in shear force trends with regard to occupant orientation lies in the ATD models incorporated during testing. The former study incorporated the Hybrid III ATD (Benson et al. 1996) and the latter incorporated the BioRID II (Latchford et al. 2005). Differences with regard to ATD models were also identified by Cappon et al. (2005), wherein correlations to actual injury data were very high for the RID3D and lower for the BioRID II. These findings may suggest decreased biofidelity for lower neck shear forces in the BioRID compared to other ATD models. However, the issue has not been extensively investigated. Due to incorporation of the BioRID in the EuroNCAP whiplash assessment protocol (European New Car Assessment Programme 2009), as well as its use by the Insurance Institute for Highway Safety (RCAR-IIWPG 2008), this issue deserves more detailed attention.

Varying trends with regard to seatback orientation were demonstrated by different injury metrics. Lower neck shear force increased for more vertically oriented seatback positions. The magnitude of lower neck extension moment, Nkm, and LNL agreed with this finding, although shear force was the most sensitive to changes in seatback orientation, increasing by an average of 42 percent across both impact velocities. Agreement between lower neck shear force, Nkm, and LNL may result from all three metrics incorporating shear force components, LNL in the lower neck and Nkm in the upper neck. Metrics incorporating shear force also demonstrated high correlations to actual injury data from insurance claims (Cappon et al. 2005). However, NIC and Nij demonstrated contradictory trends. Both criteria increased for the 15 degree seatback at lower velocities and decreased for the 15 degree seatback at higher velocities. Those metrics were also less capable of predicting presence and absence of cervical spine soft-tissue injuries in the PMHS study by Yoganandan, Philippens, and coauthors (Philippens et al. 2000; Yoganandan et al. 2001). NIC exceeded the threshold value and Nij was well below the injury threshold for all five tests. Due to these findings, along with the knowledge that cervical segments are more flexible in shear than other loading modalities, shear force appears to be a consistent predictor for injury risk in automotive rear impacts.

The present study was successful in demonstrating the utility of lower neck loads for predicting soft-tissue injuries in low-velocity rear impacts. Of all investigated metrics, lower neck shear force was most predictive of changes in injury risk due to seatback orientation and impact velocity in terms of peak magnitude and timing. Other lower neck metrics, including LNL and bending moment, were also sensitive to changes in injury risk, although in some cases timing of peak values was well after head restraint contact. These findings agree with previous investigations that argue for the incorporation of lower neck loads in rear-impact injury criteria (Benson et al. 1996; Cappon et al. 2005; Latchford et al. 2005; Tencer et al. 2002). With regard to prediction of actual injuries in PMHS, although LNL was not computed in the studies by Yoganandan, Philippens, and coauthors (Philippens et al. 2002; Yoganandan et al. 2001), lower neck bending moment was reported in addition to lower neck shear force. Gender-dependent extension moment thresholds reported by Mertz et al. (2003) would have predicted presence or absence of injury in three of the five PMHS incorporated in those studies (Yoganandan et al. 2000). In a fourth PMHS with injury, the reported bending moment was only 1.0 Nm below the threshold reported by Mertz et al. (2003). Therefore, although not as predictive as lower neck shear force in terms of magnitude and timing, lower neck bending moment may also have application to soft-tissue injury prediction in low-speed automotive rear impacts.

Though lower neck shear force was verified as a possible injury metric for low-velocity automotive rear impacts, results are somewhat limited in scope. Correlation between the factors listed above (i.e., head restraint characteristics and occupant-related factors) and injury potential in rear impacts should be quantified prior to lower neck shear force being accepted as an injury metric. Additionally, the present study incorporated the Hybrid III ATD because it remains the standard for testing under the United States DOT rear impact assessment protocol (NHTSA 2004), although it was not originally intended for use in low-velocity rear impacts because the neck is designed to mimic the response of a pretensed human in frontal impact and cannot recreate complex cervical spine motions during the retraction phase. Comparisons between the Hybrid III, BioRID, and RID ATDs have generally identified decreased biofidelity in the rear-impact response of the Hybrid III (Linder et al. 2000; Philippens et al. 2002). However, those comparisons were primarily based on kinematic measures. Hybrid III impact response (e.g., neck loads) was generally more biofidelic. For example, a recent study compared biofidelity of Hybrid III, BioRID II, and RID3D ATDs (Yamazaki et al. 2008). Comparing ATD to human volunteer response using cumulative variance ratio, the Hybrid III mean impact response was approximately equal to the other ATDs. In addition, for measures obtained at 150 ms during impacts conducted using a horizontal acceleration sled with stock automobile seat, conditions closest to the current test series, the Hybrid III mean impact response was actually better than the BioRID II. Therefore, use of the Hybrid III ATD to quantify lower neck loads in the present study was justified. However, due to inherent differences between ATDs, verification of lower neck shear force as an injury metric should also be performed using current versions of the BioRID and RID, and ATD-specific thresholds should be quantified.

Maximum facet joint ligament strains were highest at caudal cervical levels for all test conditions. These results are supported by prior studies reporting highest capsular strain/stretch at lower cervical levels under simulated rear-impact loading of full-body PMHS, intact head-neck complexes, isolated cervical spines, and the THUMS (Total Human Model for Safety) computational model (Cusick et al. 2001; Kitagawa et al. 2008; Pearson et al. 2004; Stemper et al. 2004a; Sundararajan et al. 2004). Peak lower cervical facet joint ligament strain magnitudes from the present study were similar to the upper limit of strain magnitudes reported in an experimental study subjecting isolated cervical spines to similar rear-impact severities (Pearson et al. 2004). Timing of peak facet joint strain values is also supported by literature, wherein facet stretch at C3-C4 through C5-C6 levels attained 50 to 70 percent of maximum values at or before the time of head restraint contact (Sundararajan et al. 2004). In most cases, facet joint stretch magnitudes from that study reached a peak or plateau shortly before or after head restraint contact. These findings serve as secondary validation of the present computational model against experimental data obtained using PMHS. Additionally, strain magnitudes from the present study were sufficient to result in capsular ligament subfailures at C5-C6 and C6-C7 spinal levels (Quinn and Winkelstein 2007), which may be responsible for nociceptive symptoms commonly reported by whiplash patients.

Facet joint ligament strain was used to estimate relative injury risks between different impact conditions. As stated in the Introduction, considerable evidence exists in clinical and experimental literature to implicate lower cervical facet joints in the injury mechanism resulting from low-speed automotive rear impacts, including nonphysiologic kinematics, presence of pain facilitating neuropeptides, similar pain distribution patterns to the most commonly reported whiplash symptoms, and actual injuries identified in PMHS following single- or multiple-application rear impacts. However, facet joint injuries may not be exclusively responsible for all symptoms associated with whiplash associated disorders (WAD). Other injury theories have been proposed that may also be associated with symptoms of WAD, including anterior cervical column injury due to local or head–neck hyperextension (Macnab 1964; Mertz and Patrick 1967; Panjabi et al. 2004; Severy et al. 1955; Stemper et al. 2006b), nerve root injury due to pressure gradients in the spinal canal

(Aldman, 1986; Svensson et al. 1993), or neck muscle injury due to eccentric contraction (Brault et al. 2000; Tencer et al. 2001).

Lower neck shear force was shown to be a valid predictor for injury risk in low-velocity rear impacts. Present results and data in literature have identified higher sensitivity of lower neck shear force to changes in impact velocity and seatback orientation than other injury criteria. Those results correlated with localized spinal kinematic differences (e.g., ligament strains) obtained using a comprehensively validated computational model. To date, quantification of lower neck shear force in experimental rear-impact testing has been limited. However, based on results of the present study, further testing incorporating this metric is required. In particular, investigation of the sensitivity of lower neck shear force to other factors known to affect injury risk (i.e., gender) and verification of shear thresholds with actual injuries sustained by PMHS in experimental rear impacts are essential to the acceptance of this metric as a predictor of injury risk during low-velocity automotive rear impacts. Additionally, further correlation of real-world whiplash injury data with experimental test results obtained using up-to-date ATDs is important and warranted.



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## *REFERENCES*

Aldman B. An analytical approach to the impact biomechanics of head and neck injury. *Annu Proc Assoc Adv Automot Med.* 1986;30:439–454. October 6–8, 1986; Montreal, Quebec.

Benson BR, Smith GC, Kent RW, et al. Effect of seat stiffness in out-of-position occupant response in rear-end collisions. Paper presented at: 40th Stapp Car Crash Conference; November 4–6, 1996; Albuquerque, New Mexico.

Bostrom O, Svensson M, Aldman B, et al. A new neck injury criterion candidate based on injury findings in the cervical spinal ganglia after experimental sagittal whiplash. Paper presented at: International IRCOBI Conference on the Biomechanics of Impact; September 11–13, 1996; Dublin, Ireland.

Brault J, Siegmund G, Wheeler J. Cervical muscle response during whiplash: evidence of a lengthening muscle contraction. *Clin Biomech*. 2000;15:426–435.

Cappon H, Hell W, Hoschopf H, et al. Correlation of accident statistics to whiplash performance parameters using the RID3D and BioRID dummy. Paper presented at: International IRCOBI Conference on the Biomechanics of Impact; September 21–23, 2005; Prague, Czech Republic.

Cusick JF, Pintar FA, Yoganandan N. Whiplash syndrome: kinematic factors influencing pain patterns. *Spine*. 2001;26:1252–1258.

Dehner C, Elbel M, Schick S, et al. Risk of injury of the cervical spine in sled tests in female volunteers. *Clin Biomech.* 2007;22:615–622.

Deng B, Begeman PC, Yang KH, et al. Kinematics of human cadaver cervical spine during low speed rear-end impacts. *Stapp Car Crash J.* 2000;44:171–188.

European New Car Assessment Programme. *The Dynamic Assessment of Car Seats for Neck Injury Protection Testing Protocol.* Version 2.8. Available at: http://www.euroncap.com (accessed March 2009).

Heitplatz F, Sferco R, Fay P, et al. An evaluation of existing and proposed injury criteria with various dummies to determine their ability to predict the levels of soft tissue neck injury seen in real world accidents. Paper presented at: the 18th Technical Conference on the Enhanced Safety of Vehicles; May 19–22, 2003; Nagoya, Japan.

Kitagawa Y, Yasuki T, Hasegawa J. Research study on neck injury lessening with active head restraint using human body FE model. *Traffic Inj Prev.* 2008;9:574–582.

Klinich KD, Saul RA, Auguste G, et al. *Techniques for Developing Child Dummy Protection Reference Values*. National Highway Traffic Safety Administration; 1996. Docket submission #74-14 Notice 97 Item 069.

Latchford J, Chirwa EC, Mao M. The relationship of seat backrest angle and neck injury in low-velocity rear impacts. *J Automobile Eng.* 2005;219:1293–1302.

Linder A, Bergman U, Svensson M, et al. Evaluation of the BioRID P3 and the Hybrid III in pendulum impacts to the back—a comparison to human subject test data. *Annu Proc Assoc Adv Automot Med.* 2000;44:283–297. October 2–4, 2000; Chicago, Illinois.

Linder A, Olsson T, Truedsson N, et al. Dynamic performances of different seat designs for low to medium velocity rear impact. *Annu Proc Assoc Adv Automot Med.* 2001;45:187–201. September 24–26, 2001; San Antonio, Texas.

Macnab I. Acceleration injuries of the cervical spine. *J Bone Joint Surg Am.* 1964;46-A(8):1797–1800.

Mertz HJ, Irwin AL, Prasad P. Biomechanical and scaling bases for frontal and side impact injury assessment reference values. *Stapp Car Crash J.* 2003;47:155–188.

Mertz HJ, Patrick LM. Investigation of the kinematics and kinetics of whiplash. Paper presented at: the 11th Stapp Car Crash Conference; October 10–11, 1967; Anaheim, California.

Mertz HJ, Prasad P. Improved neck injury risk curves for tension and extension moment measurements of crash dummies. *Stapp Car Crash J.* 2000;44:59–75.

Myklebust JB, Pintar F, Yoganandan N, et al. Tensile strength of spinal ligaments. *Spine*. 1988;13:526–531.

National Highway Traffic Safety Administration. Standard No. 202a; Head restraints; Code of Federal Regulations Title 49, Volume 6, Chapter 5, Part 571.202a. Available at: http://www.nhtsa.gov (accessed May 2009).

Ono K, Kaneoka K, Wittek A, et al. Cervical injury mechanism based on the analysis of human cervical vertebral motion and head-neck-torso kinematics during low speed rear impacts. Paper presented at: 41st Stapp Car Crash Conference; November 13–14, 1997; Lake Buena Vista, Florida.

Panjabi MM, Ito S, Pearson AM, et al. Injury mechanisms of the cervical intervertebral disc during simulated whiplash. *Spine.* 2004;29:1217–1225.

Pearson AM, Ivancic PC, Ito S, et al. Facet joint kinematics and injury mechanisms during simulated whiplash. *Spine.* 2004;29:390–397.

Philippens M, Cappon H, van Ratingen M, et al. Comparison of the rear impact biofidelity of BioRID II and RID2. *Stapp Car Crash J.* 2002;46:461–476. September 20–22, 2000; Montpellier, France.

Philippens M, Wismans J, Cappon J, et al. Whole body kinematics using post mortem human subjects in experimental rear impact. Paper presented at: the International IRCOBI Conference on the Biomechanics of Impact; 2000.

Pramudita JA, Ono K, Ejima S, et al. Head/neck/torso behavior and cervical vertebral motion of human volunteers during low speed rear impact: mini-sled tests with mass production car seats. Paper presented at: the International IRCOBI Conference on the Biomechanics of Impact; September 19–21, 2007; Maastricht, The Netherlands.

Prasad P, Daniel RP. A biomechanical analysis of head, neck, and torso injuries to child surrogates due to sudden torso acceleration. Paper presented at: the 28th Annual Stapp Car Crash Conference; November 6–7, 1984; Chicago, Illinois.

Quinn KP, Winkelstein BA. Cervical facet capsular ligament yield defines the threshold for injury and persistent joint-mediated neck pain. *J Biomech*. 2007;40:2299–2306.

RCAR-IIWPG. *RCAR-IIWPG Seat/Head Restraint Evaluation Protocol.* Version 3. Insurance Institute for Highway Safety; 2008. Available at: http://www.iihs.org. (accessed January 2009).

Schmitt KU, Muser MH, Walz FH, et al. Nkm—a proposal for a neck protection criterion for low-speed rear-end impacts. *Traffic Inj Prev.* 2002;3:117–126.

Severy DM, Mathewson JH, Bechtol CO. Controlled automobile rear-end collisions, an investigation of related engineering and medical phenomena. *Can Serv Med J*. 1955;11:727–759.

Society of Automotive Engineers. *Instrumentation for Impact Test—Part 1—Electonic Instrumentation-SAE J211/1.* 1995. Warrendale, Pennsylvania.

Stemper BD, Yoganandan N, Gennarelli TA, et al. Toward a lower neck rear impact injury criterion—correlation of lower neck loads with spinal kinematics. Paper presented at: the International IRCOBI Conference on the Biomechanics of Impact; September 19–21, 2007; Maastricht, The Netherlands.

Stemper BD, Yoganandan N, Pintar FA, et al. Anterior longitudinal ligament injuries in whiplash may lead to cervical instability. *Med Eng Phys.* 2006b;28:515–524.

Stemper BD, Yoganandan N, Pintar FA. Effect of head restraint backset on head-neck kinematics in whiplash. *Accid Anal Prev*. 2006a;38:317–323.

Stemper BD, Yoganandan N, Pintar FA. Effects of abnormal posture on capsular ligament elongations in a computational model subjected to whiplash loading. *J Biomech.* 2005;38:1313–1323.

Stemper BD, Yoganandan N, Pintar FA. Gender-and region-dependent local facet joint kinematics in rear impact: implications in whiplash injury. *Spine.* 2004a;29:1764–1771.

Stemper BD, Yoganandan N, Pintar FA. Intervertebral rotations as a function of rear impact loading. *Biomed Sci Instrum.* 2002;38:227–231.

Stemper BD, Yoganandan N, Pintar FA. Validation of a head-neck computer model for whiplash simulation. *Med Biol Eng Comput.* 2004b;42:333–338.

Stemper BD, Yoganandan N, Rao RD, et al. Influence of thoracic ramping on whiplash kinematics. *Clin Biomech.* 2005a;20:1019–1028.

Stemper BD, Yoganandan N, Rao RD, et al. Reflex muscle contraction in the unaware occupant in whiplash injury. *Spine.* 2005b;30:2794–2798.

Sundararajan S, Prasad P, Demetropoulos CK, et al. Effect of head-neck position on cervical facet stretch of post mortem human subjects during low speed rear end impacts. *Stapp Car Crash J.* 2004;48:331–372.

Svensson MY, Aldman B, Hansson HA, et al. Pressure effects in the spinal canal during whiplash extension motion: a possible cause of injury to the cervical spinal ganglia. Paper presented at: the International IRCOBI Conference on the Biomechanics of Impact; September 8–10, 1993; Eindhoven, The Netherlands.

Szabo TJ, Voss DP, Welcher JB. Influence of seat foam and geometrical properties on BioRID P3 kinematic response to rear impacts. *Traffic Inj Prev.* 2003;4:315–323.

Szabo TJ, Welcher JB. Human subject kinematics and electromyographic activity during low speed rear impacts. Paper presented at: the 40th Stapp Car Crash Conference; November 4–6, 1996; Albuquerque, New Mexico.

Takeshima T, Omokawa S, Takaoka T, et al. Sagittal alignment of cervical flexion and extension: lateral radiographic analysis. *Spine.* 2002;27:E348–E355.

Tencer AF, Mirza S, and Bensel K. Internal loads in the cervical spine during motor vehicle rear-end impacts: the effect of acceleration and head-to-head restraint proximity. *Spine.* 2002;27:34–42.

Tencer AF, Mirza S, Bensel K. The response of human volunteers to rear-end impacts: the effect of head restraint properties. *Spine.* 2001;26:2432–2440.

Winkelstein BA, Nightingale RW, Richardson WJ, et al. The cervical facet capsule and its role in whiplash injury: a biomechanical investigation. *Spine.* 2000;25:1238–1246.

Yamazaki K, Ono K, Ishii M. Biofidelity of rear impact dummies in low-speed rear-end impact—comparison of rigid seat and mass-production car seat with human volunteers. Paper presented at: the International IRCOBI Conference on the Biomechanics of Impact; September 17–19, 2008; Bern, Switzerland.

Yoganandan N, Cusick JF, Pintar FA, et al. Whiplash injury determination with conventional spine imaging and cryomicrotomy. *Spine.* 2001;26:2443–2448.

Yoganandan N, Pintar FA, Stemper BD, et al. Biomechanics of human occupants in simulated rear crashes: documentation of neck injuries and comparison of injury criteria. *Stapp Car Crash J.* 2000;44:189–204.