TOWARD A LOWER NECK REAR IMPACT INJURY CRITERION – CORRELATION OF LOWER NECK LOADS WITH SPINAL KINEMATICS

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ABSTRACT

Experimental research and analysis was focused on developing a lower neck rear impact injury criterion. Development was based on correlations between spinal kinematics and lower neck loads obtained using human cadaver intact head-neck complex specimens. Kinematics were significantly dependent ($p<0.05$) upon gender and spinal level. Anterior-posterior shear force demonstrated highest correlations with localized facet joint motions at C4-C5 through C6-C7 for both genders. Correlation slope was greater in females. Lower neck shear force had high predictive ability for lower cervical facet joint motions, a likely culprit for injury in low-speed automotive rear impacts. Due to gender-dependence, the injury criterion should incorporate a scaling factor for females.

Keywords: Cervical Spine, Rear Impacts, Injury Criteria, Kinematics, Biomechanics.

EARLY WHIPLASH INJURY THEORIES cited head-neck hyperextension following automotive rear impacts as the primary mechanism (Macnab 1971; Mertz and Patrick 1967; Severy *et al.* 1955). However, head restraints introduced in passenger vehicles demonstrated limited effectiveness in reducing the frequency of whiplash injuries (Kahane 1982; O'Neill *et al.* 1972; States *et al.* 1972). Since the initial purpose of automotive head restraints was to prevent head-neck hyperextension, this theory was abandoned and research was focused toward other injury mechanisms. More recent injury theories included nerve root damage due to pressure gradients in the cervical spinal canal (Aldman 1986; Svensson *et al.* 1993), anterior cervical column injuries due to localized segmental hyperextension (M. M. Panjabi *et al.* 2004; Stemper *et al.* 2006), and neck muscle injuries due to eccentric contraction (Brault *et al.* 2000). However, these theories failed to account for the specific clinical symptomatology and/or chronicity of Whiplash Associated Disorders (WAD).

A large group of studies focused on lower cervical facet joint injury, which may more accurately account for the common clinical presentation of WAD. For example, earlier studies conducted by Bogduk, Barnsley, and co-workers identified that lower cervical facet joint injury was responsible for posterior neck pain (Aprill *et al.* 1990; Barnsley *et al.* 1995; Bogduk and Marsland 1988; Dwyer *et al.* 1990), reported as the primary symptom in a majority of whiplash patients (Cassidy *et al.* 2000; Hildingsson and Toolanen 1990; Hohl 1974; Karlsborg *et al.* 1997; Mayou and Bryant 1996; Norris and Watt 1983; Radanov *et al.* 1995; Sturzenegger *et al.* 1994; Temming and Zobel 2000). Using level-specific nerve blocks, these investigators determined the manifestation of facet-joint mediated pain at all cervical levels. Other studies identified the presence of mechano-receptors/nociceptors (McLain 1994; Ohtori *et al.* 2001; Suseki *et al.* 1997), substance P (Beaman *et al.* 1993; Kallakuri *et al.* 2004), and Calcitonin Gene-Related Peptide (CGRP) (Kallakuri *et al.* 2004) in facet joints. Each of these has been associated with the pain process. For example, mechano-receptors may generate a noxious response during non-physiologic levels of joint capsule stretch and Substance P is responsible for initiation of pain while CGRP is associated with persistence of pain (Rothman *et al.* 2005). These anatomically- and physiologically-based investigations clearly identified that cervical facet joints are capable of generating the primary symptoms of WAD. In addition, biomechanics-based investigations were focused on defining the role of facet joints in the injury mechanism.

Experimental investigations have used a variety of models to delineate the biomechanical role of facet joints in whiplash. Using whole-body, intact head-neck, and isolated cervical specimens, facet joint kinematic analysis during initial whiplash stages demonstrated increased and abnormal joint motions (Cusick *et al.* 2001; Deng *et al.* 2000; Pearson *et al.* 2004; Stemper *et al.* 2004). For example, during initial whiplash phases, lower cervical facet joints demonstrate region-dependent kinematics, with uniform shear across the sagittal plane coupled with distraction in ventral and compression in dorsal joint regions (Stemper *et al.* 2004). This kinematic pattern was recently shown to be significantly different than physiologic kinematics (Stemper *et al.* 2005a) and may lead to non-physiologic joint capsule stretch magnitudes that may result in subfailure or catastrophic ligamentous failure. This theory is supported by whole-body cadaver investigations that identified lower cervical joint capsule injuries following rear-impact acceleration (Deng *et al.* 2000; Yoganandan *et al.* 2001). Recently, using a novel rodent model, cervical facet joint stretch was associated with allodynia (Lee *et al.* 2004). Although pain symptoms were manifested in the forepaw, this study was the first to associate experimentally induced facet joint injury through joint capsule stretch with noxious response. The anatomically-, physiologically-, and biomechanics-based studies outlined in this section, among others, have identified lower cervical facet joints as a key factor in the whiplash injury mechanism.

FACTORS INFLUENCING SPINAL KINEMATICS AND INJURY: Several factors have been consistently cited in literature to influence whiplash injury likelihood and/or severity. Among others, female gender (Balla 1980; Borchgrevink *et al.* 1998; Cassidy *et al.* 2000; Hohl 1974; Radanov *et al.* 1995; Spitzer *et al.* 1995; Sturzenegger *et al.* 1994), older age (Hohl 1974; Kyhlback *et al.* 2002; Maimaris *et al.* 1988; Parmar and Raymakers 1993; Radanov *et al.* 1995), higher impact severity (Atherton *et al.* 2006), abnormal posture (Hohl 1974; Norris and Watt 1983), lack of awareness (Awerbuch 1992; Hendriks *et al.* 2005; Sturzenegger *et al.* 1994), and rotated head positions (Radanov *et al.* 1995; Sturzenegger *et al.* 1994) were clinically correlated with increased risk of injury or greater persistence of neck pain following automotive rear impacts. Female gender is the factor cited most frequently as having an adverse effect on injury outcome. This factor was also experimentally shown to influence spinal kinematics during whiplash. Female intact human head-neck complex specimens demonstrated significantly increased level-by-level segmental angulations (Stemper *et al.* 2003) and localized lower cervical facet joint motions (Stemper *et al.* 2004) during initial whiplash stages. Following an automotive rear impact, the cervical spine demonstrates abnormal, level-dependent behavior characterized by flexion at superior segments (C2-C3), and increasing levels of extension at inferior segments (C4-C5 to C6-C7) (Stemper *et al.* 2003). Increased spinal motions in females, particularly in lower cervical facet joints, leads to greater ligamentous stretch and an increased likelihood of soft-tissue injury for that population. Although the automotive safety community has indirectly acknowledged the influence of gender through the development of anthropometry-specific test dummies (e.g., fifth percentile female), scaling factors for gender have not been included in rear-impact injury criteria, used to assess the likelihood of injury in specific impacts and the ability of automotive seats to protect an occupant.

PREVIOUS INJURY CRITERIA: Rear impact injury tolerance criteria were developed to assess injury risk in specific impacts. These criteria are typically used in conjunction with ATDs to assess safety of automobile seats, head restraints, and other enhancements. Previously developed tolerance criteria assess injury risk based on either head-neck-thorax kinematics or neck loads (forces and moments). Kinematic tolerance criteria include NIC (Bostrom *et al.* 1996), NDC (Viano and Davidsson 2002; Viano and Olsen 2001), and IV-NIC (M. Panjabi *et al.* 1999; M. M. Panjabi *et al.* 2005). NIC was based the theory that spinal nerve root injury may occur due to pressure gradients in the spinal canal resulting from head-neck extension or retraction (Aldman 1986), and is computed using relative acceleration and velocity between the head and thorax. IV-NIC compares intervertebral motions sustained during rear impact to physiologic levels, although it was not validated and physiologic limits were not identified (Schmitt *et al.* 2002). NDC couples maximum occipital condyle (OC)

rotation with maximum OC to T1 anterior-posterior and vertical displacement. However, the primary limitation of NDC, as well as all kinematic tolerance criteria, is that contemporary ATDs demonstrate limited kinematic biofidelity. For example, ATDs are not capable of reproducing accurate level-by-level segmental angulations or head retraction relative to the thorax. A large body of evidence indicates that the retraction phase may be the most injurious period following automotive rear impact. Due to accurate mass and inertial properties of body segments, ATDs are more capable of reproducing biofidelic neck loads. Therefore, the present research will develop a new rear impact tolerance criterion based on neck loads instead of occupant kinematics.

Tolerance criteria based on neck loads include Nij, originally developed by NHTSA for frontal impact (Kleinberger *et al.* 1999; Klinich *et al.* 1996; Prasad and Daniel 1984), and its derivatives such as Nkm (Schmitt *et al.* 2002) and Nte (Kleinberger *et al.* 1999). Although not a derivative of Nij, the LNL criterion evaluates rear impacts using a similar computation (Heitplatz *et al.* 2003). The basic principle behind these criteria is to couple uniaxial neck forces and moments and compare maximum values to previously determined critical values indicative of physiologic limits. The Nij-derived criteria couple varying combinations of upper-neck forces and moments: Nij (Nte) couples extension moment with tensile force and Nkm couples extension moment with shear force. The major limitation of these criteria is that upper-neck loads are distant from the point of load transfer to the occupant and the cervical region most commonly injured. As stated above, rear impact acceleration is transferred to the thorax through the seatback and injuries most commonly affect lower cervical levels. The LNL criterion compares lower neck moments (extension and lateral bending) and tri-axial forces (anterior-posterior and lateral shear, and tension/compression) to critical intercept values. While measuring loads in the proper position (lower neck), the computation of this metric is overly complex and includes loading components that likely have a minimal effect on whiplash injury likelihood. In addition, due to the incorporation of lower neck moments, LNL values are highly dependent upon specific ATDs due to differences in neck geometry and lack of a temporal window. Maximum LNL values may occur during the extension phase, when extension moments attain maximum values. A limitation common to all previously developed injury criteria is a lack of dynamic kinematic validation relative to localized spinal kinematics. A second limitation is the lack of a scaling factor for at-risk populations such as females. Therefore, the present research was focused on development of a new rear impact injury tolerance criterion based on lower neck loads and validated against localized spinal kinematics obtained during rear-impact acceleration.

PRESENT FOCUS: The purpose of this analysis was to correlate lower neck loads with spinal kinematics from rear impact experiments previously conducted at our laboratory. Lower neck forces and moments directly measure driving loads for head-neck kinematics in rear impact, transmitted to the thorax through the seatback. This analysis will lead to development of a lower neck rear impact injury criterion. A unique aspect of the study is correlation of lower neck loads with localized facet joint kinematics, the spinal structures most likely injured during rear impacts. Due to gender differences in the rate of injury and spinal kinematics following rear impact, incorporation of a scaling factor was investigated.

MATERIALS AND METHODS

Experiments were conducted previously and are summarized here (Stemper *et al.* 2004; Stemper *et al.* 2003). Fresh intact human cadaver head-neck complexes were screened for HIV, and hepatitis A, B, and C using standard guidelines. Neutral and extension radiographs were obtained to rule out pre-existing spinal trauma or substantial spinal degeneration. Specimens were isolated at the level of T2, and consisted of the head, ligamentous cervical spine, and first and second thoracic vertebrae with intact skin and musculature.

Head-neck complexes were fixed using polymethylmethacrylate (PMMA) at the level of T1, which was given a 25-deg anterior orientation to facilitate normal sagittal plane lordosis. An approximately 5 cm wide portion of skin and musculature was removed from the right lateral side of the neck to facilitate reflective target placement. Targets were placed in the

lateral side of the vertebral body and lateral mass of each vertebra from C4 to C7 to track angular kinematics. Smaller targets were used to outline cervical facet joints from C4-C5 to C6-C7 levels. Two targets were placed on the superior facet process, and two were placed on the inferior process at ventral and dorsal joint corners (Fig. 1).

SPECIMEN TESTING AND MOTION ANALYSIS: Specimens were rigidly attached to a minisled apparatus and oriented with the T1 PLL directly superior to the center of the load cell. To simulate normal driving position, the Frankfort plane (line joining the auditory meatus and inferior edge of the orbit) was held horizontal and occipital condyles were aligned directly superior to the T1 vertebral body. Rear impact was applied using a pendulum and measured with a linear accelerometer mounted on the minisled. Input pulses were classified as change in velocity of T1, and each specimen was exposed to four severities: 0.6, 1.3, 1.8, and 2.6 m/sec. Mean accelerations associated with these changes in velocity were 0.73±0.06, 1.34±0.10, 1.94±0.10, and 2.32±0.12 g's. Peak accelerations associated with these changes in velocity were 1.26±0.12, 2.22±0.17, 3.14±0.18, and 3.78±0.20 g's. The test matrix was such that two tests were conducted at 0.6 m/sec to obtain baseline kinematics. These low-velocity tests were repeated once between each higher input velocity. During analysis, kinematics from low-velocity tests (0.6 m/sec) were used to assess soft-tissue failure that may have been sustained during the previous higher velocity test (1.3, 1.8, or 2.6 m/sec). Specimens were also investigated for failure after each higher velocity test using visual inspection and radiography. Injury was defined as dislocation of the facet or disc joint, ligament, endplate failure, or body disruption. Testing stopped if injury was detected.

Spinal kinematics were recorded at 1,000 frames per second from the right lateral side using a high-resolution digital imaging system (Redlake MASD Inc., San Diego, CA). Temporal angular motions were defined for each vertebra from C4 through C7. Segmental angles were computed as the sagittal plane angle of one vertebra relative to the adjacent vertebra (C4-C5 to C6-C7) and filtered according to SAE specifications using a Class 60 digital filter. Linear facet joint motion was analyzed in a joint-specific localized anatomic coordinate system, with the origin at the dorsal inferior target. Shear motion was defined as motion of the superior target along the local x-axis with respect to the inferior target. Distraction was defined as motion of the superior target along the local z-axis with respect to the inferior target (distraction was positive, compression was negative). Resultant facet joint motion was computed in ventral and dorsal anatomic joint regions at C4-C5 thought C6-C7 levels.

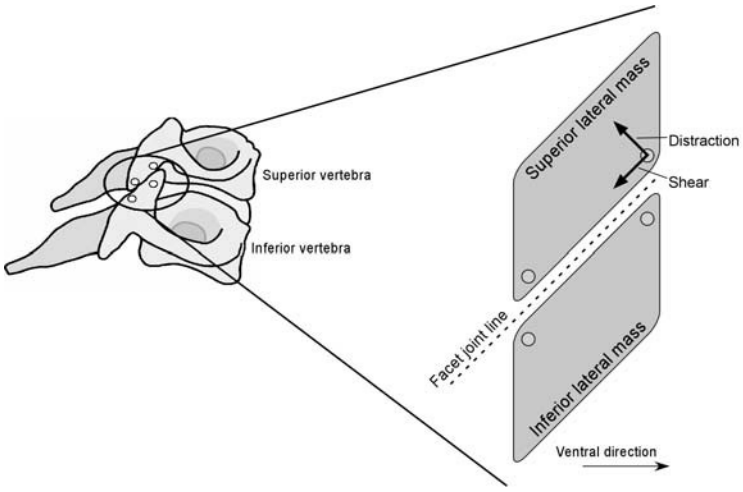

Fig. 1 – Localized facet joint kinematics.

Loading at the cervical spine base was measured at 12,500 Hz using a six-axis load cell attached to the minisled. The load cell coordinate system measured posterior-to-anterior force in the positive x-axis, right-to-left lateral force in the positive y-axis, and inferior-to-superior force in the positive z-axis. Right lateral bending (x-axis), flexion (y-axis), and left axial rotation (z-axis) were positive moments. Kinetics data were filtered according to SAE

specifications. Lower neck axial and shear forces were filtered with an SAE Class 1,000 digital filter, and lower neck moments were filtered with an SAE class 600 digital filter.

DATA REDUCTION: Cervical S-curvature was delineated as the time at which the spine experienced non-physiologic curvature, with flexion at upper segments and extension at lower segments. Segmental angulations (flexion/extension) and resultant facet joint motions in dorsal and ventral joint regions were computed during maximum S-curvature, defined as the time at which maximum flexion occurred in the C2-C3 segment. Mean time of attainment was computed at each velocity.

Three factor analysis of variance (ANOVA) was used to identify significant differences ($p<0.05$) in kinematic and kinetic data based on gender, impact velocity, and spinal level. Data from 1.2, 1.8., and 2.6 m/sec tests were analyzed. Spinal kinematics were correlated to lower neck loads at the time of maximum S-curvature using simple regression. All three impact severities were grouped for this analysis. Strength of correlation was assessed using R-squared coefficients. Correlation slope was used to correlate lower neck loading magnitudes to spinal kinematics. Slope magnitudes were used to extrapolate facet joint motions to ligament failure thresholds defined in a previous investigation (Myklebust *et al.* 1988).

RESULTS

Four male (62±15 years) and five female (55±17 years) head-neck complexes were exercised in rear impact and soft-tissue injuries were not identified. Maximum S-curvature occurred earlier with increasing impact severity (Table 1). Time of peak shear force (Fx) increased while time of peak extension moment (My) decreased with impact severity (Table 1). Time of peak axial force (Fz) was not affected by impact severity. Time of peak shear force occurred prior to maximum S-curvature. Mean shear force and extension moment magnitudes obtained at the time of maximum S-curvature significantly increased with increasing velocity and were significantly lower in female specimens. Mean axial force was greater in females. Mean segmental angulations and facet joint kinematics, separated by gender, were presented previously (Stemper *et al.* 2004; Stemper *et al.* 2003).

Table 1. Timing of kinematic and kinetic events (msec).

	Peak S-curvature	Peak Fx	Peak My	Peak Fz
1.2 m/sec	80±19	42±9	189±33	89±19
1.8 m/sec	77±15	49±8	155±27	90±33
2.6 m/sec	76±17	51±8	140±24	83±20

ANOVA analysis demonstrated resultant facet joint motions in ventral and dorsal joint regions were significantly dependent upon gender (Table 2). Ventral facet joint motions were significantly dependent upon impact severity. A significant interaction between gender and spinal level was evident for facet joint motions in both anatomic regions. In other words, gender and spinal level had confounding effects on facet joint motions. Segmental angulations were significantly dependent upon gender and spinal level. Due to the influence of gender and spinal level, correlations between spinal kinematics and lower neck loads were independently generated for both genders and the three investigated lower cervical levels.

Table 2. ANOVA results for kinematics data (p-values).

	Ventral FJ Motions	Dorsal FJ Motions	Segmental Angles
Gender (G)	0.004	0.011	0.033
Spinal Level (SL)	0.771	0.339	0.022
Impact Severity (IS)	0.034	0.074	0.188
G * SL	0.039	0.031	0.879
G * IS	0.902	0.312	0.920
SL * IS	0.774	0.815	0.384
G * SL * IS	0.670	0.547	0.818

In general, lower neck loads demonstrated higher correlation with ventral facet joint kinematics than dorsal facet joint kinematics and segmental angulations (Table 3). Average R-squared values of lower neck loads with ventral facet joint kinematics across the three spinal levels was 0.80, compared to 0.74 and 0.72 for dorsal facet joint kinematics and segmental angulations, respectively. Correlation slope increased at inferior levels for all kinematic measures in males and for segmental angulations in females. However, linear correlation slope decreased at inferior levels for facet joint motions in females. Overall, facet joint motions and segmental angulations correlated best with shear force. Ventral facet joint motions demonstrated the highest correlation to shear force (0.88), without regard to gender and spinal level. Axial force demonstrated the weakest correlation with spinal kinematics.

Table 3. R-squared values for linear correlations.

	Male								
	C4-C5			C5-C6			C6-C7		
	VFJ	DFJ	SA	VFJ	DFJ	SA	VFJ	DFJ	SA
Fx	0.94	0.88	0.80	0.78	0.72	0.71	0.93	0.92	0.96
My	0.91	0.89	0.57	0.66	0.59	0.52	0.91	0.82	0.78
Fz	0.60	0.52	0.70	0.24	0.20	0.20	0.61	0.49	0.68
	Female								
	C4-C5			C5-C6			C6-C7		
	VFJ	DFJ	SA	VFJ	DFJ	SA	VFJ	DFJ	SA
Fx	0.92	0.91	0.76	0.89	0.75	0.80	0.82	0.79	0.94
My	0.92	0.91	0.73	0.95	0.86	0.72	0.88	0.81	0.93
Fz	0.89	0.90	0.76	0.76	0.67	0.72	0.79	0.75	0.75

* VFJ: Ventral facet joint motion; DFJ: Dorsal facet joint motion; SA: Segmental angulations

EXTRAPOLATION TO FAILURE THRESHOLDS: As stated above, specimens did not sustain ligamentous failure under the present protocol. Therefore, linear correlations developed in the present analysis were extrapolated to failure levels using slope of the linear correlation (Table 4).

Table 4. Slope values for linear correlations.

	Male								
	C4-C5			C5-C6			C6-C7		
	VFJ	DFJ	SA	VFJ	DFJ	SA	VFJ	DFJ	SA
Fx	0.009	0.009	0.029	0.011	0.011	0.035	0.012	0.013	0.039
My	0.059	0.060	0.187	0.071	0.070	0.237	0.085	0.088	0.281
Fz	0.032	0.031	0.116	0.029	0.027	0.080	0.045	0.044	0.143
	Female								
	C4-C5			C5-C6			C6-C7		
	VFJ	DFJ	SA	VFJ	DFJ	SA	VFJ	DFJ	SA
Fx	0.019	0.018	0.038	0.014	0.013	0.049	0.014	0.014	0.064
My	0.155	0.151	0.324	0.103	0.080	0.373	0.097	0.096	0.537
Fz	0.049	0.049	0.101	0.036	0.033	0.125	0.038	0.038	0.158

* VFJ: Ventral facet joint motion; DFJ: Dorsal facet joint motion; SA: Segmental angulations

A previous investigation at our laboratory outlined catastrophic tensile failure thresholds for isolated spinal ligaments (Myklebust *et al.* 1988). Facet joint capsules sustained 9.1, 8.7, and 10.0 mm of distraction for catastrophic failure at C4-C5, C5-C6, and C6-C7. Joint capsules typically do not sustain catastrophic-type failures in low speed rear impacts. A recent study reported that ligamentous yield occurred at approximately 80% of catastrophic failure levels and represented the threshold for initiation of pain (Quinn and Winkelstein in press). Therefore, combining results from these two studies, distraction magnitudes required for ligamentous yield are 7.3, 7.0, and 8.0 mm at C4-C5, C5-C6, and C6-C7. These distraction magnitudes can be multiplied by slope values in Table 4 to determine shear force, bending moment, and axial force levels associated with mechanical yield of lower cervical facet joint capsules. In the present study, shear force demonstrated the highest correlation

with spinal kinematics (Table 3). Therefore, critical shear force magnitudes associated with lower cervical joint capsule yield are presented in table 5.

Table 5. Critical values for lower neck shear (N).

Spinal level	Male shear force	Female shear force
C4-C5	811	384
C5-C6	636	500
C6-C7	667	571

Minimum values from data reported in Table 5 can be used as gender-dependent tolerance limits for lower neck shear force. As expected, female tolerance limits (384 N) were considerably lower than male tolerance limits (636 N). These limits represent the mean posterior-to-anterior shear force measured at the cervico-thoracic junction, required for subcatastrophic failure of lower cervical facet joint capsules.

DISCUSSION

The present study described a novel analysis of correlation between lower neck loads and spinal kinematics. The end goal of this research remains development of a new lower neck injury criterion validated against spinal kinematics obtained during experimental testing. In comparison to upper neck loads, lower neck loads are more appropriate to assess injury likelihood as head-neck kinematics are driven by seatback interaction with the thorax following automotive rear impact. Therefore, thorax acceleration is directly responsible for spinal motions that may exceed soft-tissue thresholds, leading to injury. The new rear impact injury criterion will incorporate lower neck load cell data relative to experimentally defined critical load levels (Table 5) to assess the likelihood of spinal soft tissue injury in specific rear impacts. This injury criterion will be the first to incorporate a scaling factor for gender, as injury risk was demonstrated to be higher in females.

CRITICAL LOAD DETERMINATION: Critical load magnitudes outlined in Table 5 were determined by extrapolating presently defined correlations to previously reported tensile yield distractions for capsular ligaments. This was required as soft-tissue injury was not sustained by any experimental specimens used for the present analysis. A majority of whiplash injuries occur at subfailure levels, wherein tissues are stretched beyond physiologic limits although gross failure does not occur. Therefore, ligamentous yield values were approximated based on previous testing and are presented in Table 5.

INJURY MECHANISM: Highest correlations were evident between shear forces and ventral facet joint kinematics, indicating that lower cervical facet joint motions predictably increase with lower neck shear forces. Because facet joint motions are directed posteriorly (the superior process translates posteriorly relative to the inferior process), shear motion coupled with distraction in ventral joint regions leads to joint capsule stretch (Stemper *et al.* 2004). This does not occur in dorsal joint regions as shear motion is coupled with compression, which may alleviate joint capsule stretch and lead to 'pinching' of the joint capsule or synovial fold (Cusick *et al.* 2001). Therefore, by correlating lower neck shear force with lower cervical ventral facet joint motions, lower neck shear force may be used to estimate joint capsule stretch. Because ligaments were shown to fail in tension (Yoganandan *et al.* 2000a), the present correlations can provide a direct measure of the likelihood of ligamentous failure.

Early theories on involvement of lower cervical facet joints in the whiplash injury mechanism hypothesized compression in the dorsal region leading to contact of opposing processes or joint capsule impingement. In 1998, Ono et al. presented an analysis of the instantaneous axis of rotation at C5-C6 in volunteers exposed to rear impacts on an inclined sled and demonstrated upward shift of the axis (Ono *et al.*). The authors posited that this may lead to joint capsule impingement. Earlier in the same year, Yoganandan et al. experimentally measured facet joint kinematics in cadaveric specimens and reported "accentuated local posterior compression" of lower cervical facet joints (Yoganandan *et al.*

1998). In the presence of synovial fold, posterior compression could lead to joint capsule pinching and nociceptive response. More recent theories have advanced facet joint shear resulting in joint capsule tension as the injury mechanism. Widening and disruption of facet joint capsules following single-cycle exposure to rear impact supports the theory of joint capsule tensile failure (Yoganandan *et al.* 2000b). Further analysis by the same group using cryomicrotomy identified facet joint compromise with capsular ligament tears in all experimental specimens (Yoganandan *et al.* 2001). In 2000, Deng et al. experimentally demonstrated joint capsule stretch in lower cervical facet joints during repeated loading of full-body cadavers (Deng *et al.*). This study dynamically confirmed a facet joint shear theory advanced by the same group in 1996 (Yang and Begeman). Cusick et al. hypothesized that experimentally measured joint capsule tension in upper cervical facet joints could lead to neuropathic pain (Cusick *et al.* 2001). It was later reported by Stemper et al. that facet joint motions coupled with eight degrees of segmental extension during whiplash were significantly increased relative to facet joint motions coupled with the same magnitude of physiologic segmental extension (Stemper *et al.* 2005a). The authors cited presence of a large posterior-to-anterior shear component as responsible for increased capsular ligament stretch during whiplash as compared to physiologic extension. Recently, clinical and experimental reports have identified a gender gap in whiplash injury susceptibility. In 2004, Stemper et al. reported lower cervical joint capsule stretch (shear plus distraction) in anterior regions was greater in females (Stemper *et al.*), thereby supporting clinical reports of greater whiplash injury rates in females. Further research by the same group demonstrated decreased cartilage cover on opposing articular surfaces in females, which may contribute to higher rates of facet-mediated pain following rear impact (Yoganandan *et al.* 2003). In 2006, Ono et al. reported greater facet joint shear strain in females (Ono *et al.*). The authors hypothesized the possibility of injury risk estimation based on the "state of discomfort of volunteers after the experiments and the strain of intervertebral disc and facet joints." The current analysis was aimed at development of a lower cervical rear impact injury criterion based on the facet joint shear/tension injury mechanism and recognizing significant biomechanical differences between males and females.

Peak spinal motions occurred subsequent to the time of maximum S-curvature. However, spinal kinematics and lower neck loads were quantified during this time for two reasons. A head restraint was not included in the current experimental setup. These devices are included in all passenger vehicles in the United States. Therefore, the head was allowed to extend beyond the point that a head restraint would permit during an actual automotive rear impact. More importantly, whiplash injuries occur during the retraction phase, when the head is translating posteriorly relative to the thorax with minimal extension rotation (Penning 1992). Retraction of the head relative to the thorax results in S-shaped spinal curvature, with maximum S-curvature occurring at approximately the same time as maximum retraction. Therefore, spinal kinematics and lower neck loads were measured at approximately the time that whiplash injuries are likely to occur.

INFLUENCING FACTORS: Occupant- and crash-related factors were shown to influence outcome following automotive rear impacts. In the present analysis, only gender was incorporated in the lower neck injury criterion. This decision was made because gender is the most commonly cited factor influencing injury, was experimentally shown to influence spinal kinematics during rear impact, and is accounted for in safety testing in the form of the 5th percentile female anthropomorphic test dummy. However, other factors may have a considerable affect on outcomes. For instance, older age is associated with greater injury likelihood and more persistent symptoms. This factor was also experimentally identified as an influencing factor in static and quasi-static spinal biomechanics (Board *et al.* 2006; Pintar *et al.* 1998). Therefore, it is hypothesized that age influences spinal kinematics in automotive rear impact. However, it is unknown whether age affects the relationship between lower neck loads and spinal kinematics. Mean specimen age in this analysis was 58 years. Therefore, continued rear impact testing using younger and older specimens may be required to develop a more universal injury criterion.

DISCUSSION OF THE EXPERIMENTAL MODEL: Present experimental data were derived from human cadaver intact head-neck complex specimens subjected to incremental rear impacts applied using a pendulum and minisled apparatus. The goal of this research was development of a new lower neck rear impact injury criterion to be used in conjunction with existing ATDs to assess automotive safety. As such, during experimentation a six-axis load cell was placed in the approximate position relative to the head and cervical spine as existing ATDs, including the Hybrid III and BioRID. Additionally, isolation of the head-neck complex permitted direct measurement of localized spinal kinematics (i.e., facet joints) using high speed and high resolution digital videography. Placement of the load cell and direct measurement of spinal kinematics would not have been possible using full-body cadavers or volunteers. The intact head-neck model strictly controlled the input pulse at the level of the cervico-thoracic junction, eliminating effects of thoracic ramping and permitted precise analysis of velocity-dependent mechanisms. Previous investigations of rear impact biomechanics have reported varying magnitudes of thoracic ramping, resulting from straightening of thoracic kyphosis and characterized by superior translation and posterior rotation of the upper thorax (Davidsson *et al.* 1998; Deng *et al.* 2000; McConnell *et al.* 1993; Ono *et al.* 1998). The ramping phenomenon may have been accentuated by use of an inclined sled in some studies. Additionally, effects of thoracic ramping were shown in a computational study to have a minimal influence on cervical segmental kinematics during simulated rear impacts (Stemper *et al.* 2005b). Therefore, due to incorporation of an accurately placed load cell and precise measurement of localized spinal kinematics, the present model was appropriate for development of a new lower neck rear impact injury criterion.

RECOMMENDATIONS AND FUTURE WORK: The present analysis provided gender- and level-specific correlations of lower neck loads (shear and axial forces, extension moment) to segmental and localized spinal kinematics. The logical follow-up to this analysis is development of a new lower neck rear impact injury criterion. For this to be accomplished, a number of discrete steps must be taken. Firstly, the threshold for ligamentous subfailure should be quantitatively defined. Although previous investigations have reported some results to this end (Siegmund *et al.* 2001; Winkelstein *et al.* 2000), standard deviations were extremely high and data were not separated with regard to gender and spinal level. Secondly, the form and components of the new criterion must be determined. The new injury criterion will likely take the form of previous criteria, such as Nkm, wherein lower neck loads measured during experimental testing are directly compared to critical subfailure values to determine likelihood of injury and assess vehicle and seat safety. In addition, temporal bounds must be determined as whiplash injury occurs early in the rear impact event. Although spinal kinematics correlated best with lower neck shear force, lower neck extension moment likely plays a role in the injury mechanism and should be included in injury criterion development. Correlations with extension moment were typically greater in females. The appropriate combination of shear force, bending moment, and gender scaling factor will have to be determined through continued testing of intact head-neck complexes with properly positioned head restraints and anthropomorphic test devices during a variety of impact conditions. Finally, the newly developed lower neck injury criterion should be compared to previously developed criteria, such as NIC, Nij, Nkm, and NDC to determine which of these criteria best assess the likelihood of injury under specific rear impact conditions.

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