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ORIGINAL COMMUNICATION

The Relationship Between Lower Neck Shear Force and Facet Joint Kinematics During Automotive Rear Impacts

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A primary goal of biomechanical safety research is the definition of localized injury thresholds in terms of quantities that are repeatable and easily measureable during experimentation. Recent biomechanical experimentation using human cadavers has highlighted the role of lower cervical facet joints in the injury mechanism resulting from low-speed automotive rear impacts. The present study was conducted to correlate lower neck forces and moments with facet joint motions during simulated rear impacts in an effort to define facet joint injury tolerance thresholds that can be used to assess automobile safety. Four male and four female intact head–neck complexes were obtained from cadaveric specimens and subjected to simulated automotive rear impacts using a pendulum-minisled device. Cervical spine segmental angulations and localized facet joint kinematics were correlated to shear and axial forces, and bending moments at the cervico-thoracic junction using linear regression. R^2 coefficients indicated that spinal kinematics correlated well with lower neck shear force and bending moment. Correlation slope was steeper in female specimens, indicating greater facet joint motions for a given loading magnitude. This study demonstrated that lower neck loads can be used to predict lower cervical facet joint kinematics during automotive rear impacts. Higher correlation slope in female specimens corresponds to higher injury susceptibility in that population. Although lower neck shear force and bending moment demonstrated adequate correlation with lower cervical facet joint motions, shear force is likely the better predictor due to similarity in the timing of peak magnitudes with regard to maximum facet joint motions. Clin. Anat. 24:319–326, 2011.

Key words: biomechanics; gender; cervical spine; whiplash

INTRODUCTION

Whiplash associated disorders (WAD) most commonly result from low velocity automotive rear impacts and are characterized by somatic pain along the posterior aspect of the neck, neck stiffness during bending, and suboccipital headache (Spitzer et al., 1995). Symptoms are characterized as chronic in 15–20% of WAD patients who seek medical treatment (Radanov et al., 1995; Spitzer et al., 1995; Cote et al., 2001). Structural abnormalities of the

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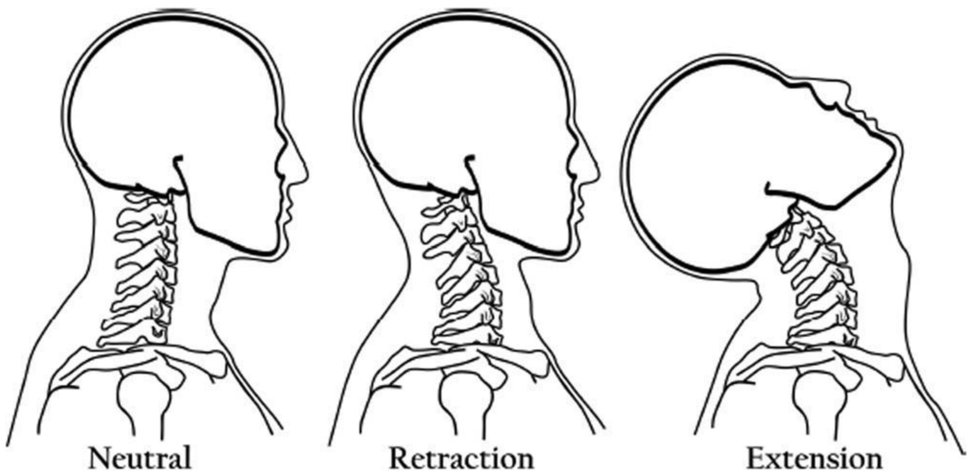

Fig. 1. Phases of head–neck kinematics following low-speed automotive rear impacts.

joints of the human neck are implicated in the chronic pain from these impacts. However, affected tissues and mechanisms of injury remain ambiguous, as injuries are often not detectable using conventional imaging modalities. Patients often see a primary care health provider and chronicity is unknown at the time of evaluation.

Human response dynamics to automotive rear impact consist of three discrete phases (Fig. 1). The first phase, known as retraction, involves anterior thrust of the seatback on the thorax, which results in forward displacement relative to the head. This initial phase leads to a non-physiologic S-shaped curvature of the cervical spine, wherein upper cervical segments experience flexion rotation and lower cervical segments experience extension rotation (Bostrom et al., 1996; Grauer et al., 1997; Kaneoka et al., 1999; Stemper et al., 2003). Inflection of the curvature is centered around the C3–C4 segment. Because of the non-physiologic cervical spine curvature, the retraction phase is often cited as the time during which soft tissue injury occurs (Penning, 1992; Bostrom et al., 1996; Ono et al., 1997; Panjabi et al., 1998; Brault et al., 2000). The cervical spine transitions into normal lordotic curvature as the head and cervical spine transition into extension following the retraction phase. Head restraint contact can occur during either the retraction or extension phase, depending on the relative position between the posterior aspect of the head and the restraint. Extension rotation magnitude is dependent on positioning of the head restraint relative to the head. The final stage begins as the head rebounds from the head restraint and the head–neck complex transitions into flexion.

Over the past 45 years, a number of theories have been proposed as the mechanism of cervical spine soft tissue injury during low velocity automotive rear impacts. The original theory involved cervical spine hyperextension leading to tension injury of the anterior longitudinal ligament and anterior intervertebral disc fibers (Macnab, 1964). A second theory posited that damage to the cervical nerve roots resulted from transient pressure elevations within the spinal canal due to relative motion between the head and thorax (Aldman, 1986; Svensson et al., 1993). Both of those injury theories were explored using animal models. The primary criticism of animal models is a lack of biofidelity relative to the human, wherein animal anthropometry and anatomical characteristics are different from humans. Experimentation using human volunteers was used to develop the theory that neck muscle injuries occur due to eccentric contraction (Brault et al., 2000). Following that logic, anterior neck muscle injuries would occur in the early stages of automotive rear impact as the head moves posteriorly, and posterior neck muscle injuries would occur during the rebound phase. However, due to ethical considerations, volunteers are required to be tested below what was considered to be the threshold of injury although 20 of 42 subjects in that study sustained acute symptoms.

Recent biomechanical experimentation using human cadavers has highlighted the role of lower cervical facet joints in the injury mechanism. Biomechanical studies offer the unique advantage of carefully examining the mechanism of injury by applying controlled and repeatable loads similar to those encountered during automobile impacts to biological models and evaluating the efficacy of these theories. Human cadavers have the advantage of incorporating the primary load-bearing anatomical structures (e.g., entire head–neck complex), and therefore, are an excellent model that can be used as an adjunct to clinical and epidemiological outcomes. An early investigation into the role of facet joints in the whiplash injury mechanism incorporated isolated cervical spines and quasi-static shear loading to hypothesize that compression forces due to thoracic ramping combined with posterior shear induces stretch of the facet joint capsule during automotive rear impact (Yang et al., 1997). In the same year, details of a dynamic head–neck complex model were published that was subsequently used to quantify kinematics of the cervical spine segments and components during simulated rear impacts (Yoganandan and Pintar, 1997; Yoganandan et al., 1998). More recent biomechanical investigations using whole-body and intact head–neck cadavers, and isolated cervical spine specimens, demonstrated non-physiologic facet joint motion during initial kinematic stages (Deng et al., 2000; Cusick et al., 2001; Pearson et al., 2004; Stemper et al., 2004; Sundararajan et al., 2004). Facet joints sustained shear motion during the retraction phase, wherein the superior articular process displaced posteriorly relative to the inferior process, coupled with tension in anterior regions and compression in posterior regions of the joint (Cusick et al., 2001; Stemper et al., 2004). This kinematic pattern was shown to be statistically significantly different from physiologic extension kinematics (Stemper et al., 2005). Injuries resulting from this mechanism include tension-induced injuries in the anterior region of the joint capsule and pinching of synovial folds or compression-induced injuries to articular cartilage or subchondral bone. A study by Yoganandan et al. (2001) confirmed facet joint and posterior column soft-tissue injuries in full-body cadavers following single exposure to simulated automotive rear impacts using a full-scale sled device. Injuries such as facet joint widening and hematoma, joint capsule disruption, and ligamentum flavum rupture were identified using cryomicrotomy. Translation of biomechanical cadaver data to clinical application is accomplished using patient studies. Pain distributions resulting from injured cervical facet joints were mapped using local anesthetic and found to result in similar pain patterns to the most commonly reported symptoms associated with WAD (Bogduk and Marsland, 1988; Aprill et al., 1990; Dwyer et al., 1990; Barnsley et al., 1995). These anatomical- and biomechanics-based studies, among others, identified lower cervical facet joints as a key factor in the whiplash injury mechanism.

A primary objective of biomechanically related injury research is quantification of tolerance thresholds in terms of measureable and repeatable quantities (Prasad and Daniel, 1984; Bostrom et al., 1996; Klinich et al., 1996; Panjabi et al., 1999; Schmitt et al., 2002; Viano and Davidsson, 2002; Heitplatz et al., 2003). These thresholds are used by safety researchers to assess the likelihood of occupant injury in specific impact scenarios. Lower neck kinetics (i.e., forces and moments) are the most likely correlates because loading is transferred from the seat to the cervico-thoracic junction during rear impacts. Therefore, the present study was conducted to correlate lower neck forces and moments with facet joint motions during simulated rear impacts. Quantification of these relationships may lead to injury tolerance thresholds that can be used to assess automobile safety.

MATERIALS AND METHODS

Data from previously published cadaveric experimentation were used in the present analysis (Stemper et al., 2003, 2004). Experimental details are summarized here. Four men (mean age: 62 ± 15 years) and four women (mean age: 49 ± 11 years) intact head–neck complexes were obtained from human cadavers and screened for HIV, and hepatitis A, B, and C using standard guidelines. Sagittal radiographs were obtained before specimen selection to rule out pre-existing spinal trauma or substantial spinal degeneration. Specimens were isolated at the level of the T2–T3 intervertebral disc, and consisted of the head, ligamentous cervical spine, and first and second thoracic vertebrae. Skin and musculature remained intact. Head–neck complexes were mounted in polymethylmethacrylate (PMMA) at the level of T1, which was given a 25° anterior orientation to facilitate normal sagittal lordosis. A small portion of skin and musculature was removed from the right side of the neck to place targets in the lateral side of the vertebral body and lateral mass of each vertebra from C2 to C7. Smaller targets were placed at the four sagittal plane corners of C4–C5 to C6–C7 facet joints. Skin and musculature were attached to the PMMA base to provide passive neck muscle resistance to head–neck motions.

Simulated automotive rear impact acceleration was applied to the specimens using a pendulum-minisled apparatus. Specimens were rigidly attached to the minisled and oriented with the T1 posterior longitudinal ligament superior to the center of the load cell. To simulate normal driving position, the Frankfort plane was maintained horizontal and occipital condyles were aligned superior to the center of the T1 vertebral body. Rear impact acceleration was applied using a pendulum, which was raised to increasing initial height for greater impact severities. The pendulum was released and allowed to swing unconstrained to strike the posterior edge of the minisled, which accelerated it horizontally along linear rails. Minisled acceleration was measured using redundant linear accelerometers. Characteristics of the acceleration versus time pulse were controlled using a spring and foam damping materials at the pendulum-minisled interface. The pulse shape was designed to match the horizontal T1 acceleration pulse from a series of full-body PMHS rear impact experiments (Yoganandan et al., 2000). Impact severity was classified as change in velocity of T1, which was computed as the integral of the acceleration versus time pulse of the minisled. Each specimen was exposed to four impact severities: 0.6, 1.3, 1.8, and 2.6 m/sec. Acceleration pulse duration was $\sim$100 msec for all impact severities and maximum acceleration was varied to achieve different velocities. The test matrix consisted of two 0.6 m/sec tests conducted prior to then again after each higher velocity test (i.e., 1.3, 1.8, and 2.6 m/sec). These tests were used to dynamically precondition the spine and to assess possible kinematic evidence of injury following testing. Specimens were investigated for tissue failure after each higher velocity test using visual inspection, palpation, and radiography. Testing stopped if injury was detected. After testing was completed, spinal kinematics were compared between lower velocity tests to identify markedly increased

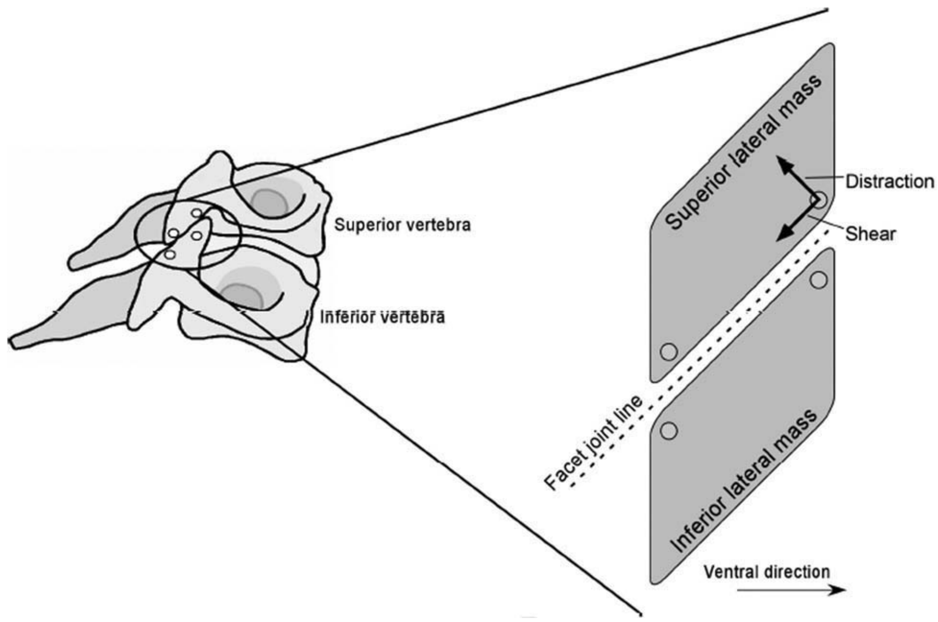

Fig. 2. Localized facet-joint coordinate system.

segmental angles at a specific level. In the event of this occurrence, injury at that level would be assumed to have occurred during the previous higher velocity test. Likewise, data from the previous and all subsequent tests using that specimen would be removed from analysis.

Axial (Fz) and shear (Fx) forces, and sagittal bending moments (My) were measured at 12,500 Hz using the load cell. Kinetics data were low-pass filtered to remove high frequency noise according to Society of Automotive Engineers (SAE) specifications. Spinal kinematics were recorded at 1,000 frames/sec (fps) from the right side using a high-resolution digital imaging system (Redlake MASD, 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. Linear facet joint motion was analyzed in a joint-specific localized anatomic coordinate system, with the origin at the dorsal inferior target (Fig. 2). Shear motion was defined as motion of the superior target along the joint line with respect to the inferior target. Distraction and compression were defined as motions perpendicular to the joint line with respect to the inferior target (distraction: positive, compression: negative). Resultant facet joint motion was computed in ventral and dorsal anatomic joint regions.

Segmental angulations and resultant facet joint motions in dorsal and ventral joint regions were computed at the time of maximum flexion at the C2–C3 segment. This time was chosen as the C2–C3 level consistently demonstrated the highest magnitude of flexion during the retraction phase. As such, cervical S-curvature was considered to be maximum at that time. Mean time of attainment was computed for each velocity. Data from 1.3, 1.8, and 2.6 m/sec tests were analyzed. Single factor Analysis of Variance (ANOVA) was used to determine statistically significant differences ($P < 0.05$) in segmental angulations, facet joint motions, axial and shear forces, and bending moments with regard to gender. Segmental angulations and facet joint motions were correlated to lower neck loads at the time of maximum S-curvature using linear regression. All three impact severities were grouped for this analysis. Strength of correlation was assessed using R^2 coefficients. Correlation slope defined the relationship between lower neck loading magnitudes and spinal kinematics.

RESULTS

All specimens were subjected to the entire experimental protocol without sustaining soft-tissue injury detected during palpation or radiography. Following the initiation of rear impact acceleration, all specimens demonstrated retraction and extension kinematic phases as identified previously. The mean time of maximum S-curvature was 80, 77, and 76 msec

TABLE 1. Lower Neck Loads and Spinal Kinematics (Mean ± Standard Deviation)

	Fx (N)	Fz (N)	My (Nm)	SA (deg)	VFJ (mm)	DFJ (mm)
Male	145.9 ± 47.8	26.2 ± 21.3	18.4 ± 11.5	5.0 ± 2.5	1.6 ± 0.8	1.6 ± 0.9
Female	126.3 ± 44.1	41.9 ± 25.7	15.4 ± 5.2	6.9 ± 3.3	1.9 ± 0.7	1.8 ± 0.8

TABLE 2. R^2 Values for Correlation of Lower Neck Loads and Spinal Kinematics

	Female			Male		
	VFJ	DFJ	SA	VFJ	DFJ	SA
Fx	0.88	0.82	0.80	0.86	0.81	0.81
My	0.87	0.81	0.78	0.80	0.73	0.61
Fz	0.80	0.75	0.70	0.45	0.37	0.47

VFJ, ventral facet joint motion; DFJ, dorsal facet joint motion; SA, segmental angulations.

for 1.3, 1.8, and 2.6 m/sec tests. Mean segmental angulations were significantly greater ($P < 0.05$) in women (Table 1). Gender differences in resultant facet joint motions in the ventral joint region approached statistical significance ($P = 0.06$), with motions in female specimens marginally greater than in males. Resultant facet joint motions in the dorsal joint region were not significantly different based on gender ($P = 0.40$). Mean shear force and bending moment magnitudes during maximum S-curvature were lower in female specimens (Table 1). Mean axial force (Fz) was greater in females. None of the three kinetic measures varied significantly ($P < 0.05$) by gender.

Lower neck loads correlated well with lower cervical segmental angulations and facet joint motions in ventral and dorsal joint regions. In general, lower neck loads demonstrated higher correlation with ventral facet joint kinematics than dorsal facet joint kinematics and segmental angulations (Table 2). Average R^2 values of lower neck loads with ventral facet joint kinematics was 0.78, compared to 0.72 and 0.70 for dorsal facet joint kinematics and segmental angulations, respectively. Average R^2 value was greatest for ventral facet joint kinematics correlation with lower neck shear force (mean $R^2 = 0.87$). Linear correlation slope was greater in female specimens than male specimens for all nine correlations (Table 3). Cervical spine kinematics correlations with lower neck bending moment were most sensitive to effects of gender, with slopes increasing an average of 65% for female specimens compared to 31% and 10% for correlations with lower neck shear and axial forces, respectively.

DISCUSSION

This manuscript describes an analysis of the correlation between lower neck loads and spinal kinematics in cadaveric specimens during simulated automotive rear impacts. The end goal of this research remains development of facet joint injury thresholds based on repeatable and measureable quantities that can be used to predict the likelihood of injury in specific automotive impact events. Lower neck loads are the most likely candidates as rear impact acceleration is transferred to the head–neck complex through interaction of the body with the seatback at the cervico-thoracic junction. Highest linear correlations were evident between ventral facet joint kinematics, and shear forces and bending moments. Strong correlations indicate that lower cervical facet joint motions predictably increase with those quantities. Because facet joint motions are directed posteriorly (e.g., 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). Therefore, lower neck loads can be used to estimate the level of joint capsule stretch in a given impact situation. Because ligaments fail in tension (i.e., stretch), lower neck loads can then be used to assess the relative likelihood of soft-tissue injury in a given impact.

Slope of the correlation between lower neck loads and facet joint kinematics was greater in female specimens for all investigated correlations. This indicates that lower cervical facet joints are subjected to greater motions and a correspondingly higher injury likelihood for a given loading magnitude. Clinical reports supported this finding by identifying that females report a higher number of whiplash injuries. Although the issue has not been definitively resolved, several studies identified possible biomechanical and anatomical explanations. Increased spinal motions in females during identical rear impacts (Stemper et al., 2003, 2004) indicates that the cervical spine responds differently to rear impact based on gender. This may be due to several factors including spinal geometry, soft tissue material properties, and neck muscle contractile properties and positioning. A recent study obtained CT scans of the cervical spine in 51 young, healthy volunteers and identified a more slender cervical column in size-matched females, which may be more susceptible to bending when subjected to rear impact acceleration (Stemper et al., 2008). Another study reported similar results (Vasavada et al., 2008). In addition, neck muscles may be oriented differently in males and females, such that male muscles provide more stability to the head–neck complex. In a sample of five female and 11 male volunteers, neck moments were shown to be approximately twice as large in males (Vasavada et al., 2001). Because the magnitude of contractile force is proportional to muscle cross-sectional area, greater muscle sizes indicate a greater ability to stabilize the head–neck complex during inertial loading. Another study identified significantly increased range of motion in lower cervical motion segments in female specimens (Nightingale et al., 2007), indicating greater flexibility in the female cervical spine. Results of these studies indicate a propensity toward

TABLE 3. Slope of Linear Correlation for Lower Neck Loads and Spinal Kinematics

	Female			Male		
	VFJ	DFJ	SA	VFJ	DFJ	SA
Fx	0.014	0.013	0.050	0.011	0.011	0.034
My	0.119	0.109	0.411	0.071	0.072	0.233
Fz	0.038	0.036	0.132	0.036	0.034	0.113

VFJ, ventral facet joint motion; DFJ, dorsal facet joint motion; SA, segmental angulations.

greater motions, increased soft tissue strains, and higher injury likelihood during inertial loading in the female cervical spine during the type of loading experienced during automotive rear impacts.

The present analysis provided gender-specific correlations of lower neck loads to localized spinal kinematics. The logical follow-up to this analysis is development of lower neck loading-based tolerance thresholds. Although correlations were approximately equal for ventral facet joint motions with lower neck shear force and bending moment, shear force may be a better indicator as maximum shear force magnitudes occurred during the retraction phase, wherein bending moments peaked at the end of the extension phase. To this end, linear correlations developed during the present analysis were extrapolated to previously reported tensile failure levels (Myklebust et al., 1988). In that study, lower cervical joint capsules sustained between 8.7 and 10.0 mm of distraction prior to catastrophic failure. However, catastrophic ligamentous failure is not commonly reported following whiplash injury. In addition, in vivo experimentation has correlated pain with facet joint distraction by reporting allodynia in rats following capsular ligament subfailure (i.e., mechanical yield) (Lee et al., 2004). An investigation of human ligamentous tensile properties identified that ligamentous yield occurred at ~80% of the distraction to failure (Chazal et al., 1985). The point of ligamentous yield can be used to determine critical force values and was computed to be an average of 7.4 mm across C4–C5, C5–C6, and C6–C7 levels. According to the relationships identified in Table 3, shear force required for 7.4 mm ventral capsular ligament distraction is ~670 N in males and 530 N in females.

The extrapolated shear force magnitudes described above are not intended to represent capsular ligament injury tolerance thresholds for low velocity automobile rear impacts. As stated above, all tests were conducted below the level of injury and resulted in no soft tissue failures in any of the tested cadavers. Rather, these extrapolations represent a first estimate at a possible new rear impact injury criterion and were justified for this purpose given the linearity of the correlation between lower neck shear force and facet joint capsular stretch. However, further validation of this metric is required and accurate gender-specific tolerance levels should be determined through an iterative process involving cadaveric specimens and dynamic testing under a variety of rear impact conditions. In addition, existing dummies capable of being used for the study of automotive rear impacts have fundamental design differences. For example, the BioRID has a segmented neck and kyphotic thoracic spine whereas the Hybrid III has a solid neck component with no thoracic spine. Dummy-specific validation of lower neck loads and specific tolerance levels needs to be accomplished prior to incorporation of this injury metric for the assessment of automotive safety. This type of validation has been performed by Prasad et al. using the Hybrid III dummy and Hybrid III dummy with RID neck (Prasad et al., 1997). That study reported dummy-specific lower neck moment tolerance values. An initial attempt to validate the concept of lower neck shear force as a rear impact injury criterion was recently performed using stock automobile seats and a the Hybrid III dummy (Stemper and Storvik, 2010). That study identified increased lower neck shear forces for rear impact conditions generally associated with higher injury likelihood (i.e., higher velocity and seatback verticality), although specific tolerance thresholds in terms of shear force magnitudes were not investigated. Lower neck shear force was more sensitive to changes in those conditions than NIC, Nij, Nkm, LNL, and lower neck bending moment. Similar studies investigating other relevant occupant- and crash-related factors coupled with cadaveric studies within the injurious range should be conducted. Occupant- and crash-related factors were shown to influence outcome following automotive rear impacts. Only gender was incorporated into the present analysis. This decision was made because gender is the most commonly cited clinical factor influencing injury, gender was experimentally shown to influence spinal kinematics during rear impact, and gender can be accounted for in automobile safety testing in the form of the fifth percentile female anthropomorphic test dummy. However, other factors may also affect injury outcomes. For instance, older age is associated with anatomical changes in the cervical spine including loss of intervertebral disc height, dehydration of the nucleus, sclerosis of the vertebral body, and osteophyte formation. These anatomical changes are likely to affect tissue mechanics and the corresponding likelihood of injury. In fact, greater injury likelihood and more persistent symptoms have been associated with older age in patients with WAD (Axtelius and Soderfeldt, 2003; Suissa, 2003; Miettinen et al., 2004; Pobereskin, 2005; Sterling et al., 2005; Crouch et al., 2006). Older age was also experimentally identified as an influencing factor in static and quasi-static spinal biomechanics (Pintar et al., 1998; Board et al., 2006). 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 fully assess the effect of this important factor.

REFERENCES

Aldman B. 1986. An analytical approach to the impact biomechanics of head and neck injury. 30th Annual Conference of the Association for the Advancement of Automotive Medicine. Montreal, Quebec. p 439–454.

Aprill C, Dwyer A, Bogduk N. 1990. Cervical zygapophyseal joint pain patterns II: A clinical evaluation. Spine 15:458–461.

Axtelius BL, Soderfeldt BR. 2003. Oral disease and psychosocial risk determinants in relation to self-assessments of general health in persons with chronic whiplash-related disorders. Swed Dent J 27:185–195.

Barnsley L, Lord SM, Wallis BJ, Bogduk N. 1995. The prevalence of chronic cervical zygapophyseal joint pain after whiplash. Spine 20:20–26.

Board D, Stemper BD, Yoganandan N, Pintar FA, Shender B, Paskoff G. 2006. Biomechanics of the aging spine. Biomed Sci Instrum 42:1–6.

Bogduk N, Marsland A. 1988. The cervical zygapophysial joints as a source of neck pain. Spine 13:610–617.

Bostrom O, Svensson M, Aldman B, Hansson H, Haland Y, Seeman T, Suneson A, Saljo A, Ortengren T. 1996. A new neck injury criterion candidate-based on injury findings in the cervical spinal ganglia after experimental sagittal whiplash. Proceedings of the International Research Council on the Biomechanics of Impact (IRCOBI). Dublin, Ireland. p 123–126.

Brault J, Siegmund GP, Wheeler JB. 2000. Cervical muscle response during whiplash: Evidence of a lengthening muscle contraction. Clin Biomech 15:426–435.

Chazal J, Tanguy A, Bourges M, Gaurel G, Escande G, Guillot M, Vanneuville G. 1985. Biomechanical properties of spinal ligaments and a histological study of the supraspinal ligament in traction. J Biomech 18:167–176.

Cote P, Hogg-Johnson S, Cassidy JD, Carroll L, Frank JW. 2001. The association between neck pain intensity, physical functioning, depressive symptomatology and time-to-claim-closure after whiplash. J Clin Epidemiol 54:275–286.

Crouch R, Whitewick R, Clancy M, Wright P, Thomas P. 2006. Whiplash associated disorder: Incidence and natural history over the first month for patients presenting to a UK emergency department. Emerg Med J 23:114–118.

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

Deng B, Begeman PC, Yang KH, Tashman S, King AI. 2000. Kinematics of human cadaver cervical spine during low speed rear-end impacts. Stapp Car Crash J 44:171–188.

Dwyer A, Aprill C, Bogduk N. 1990. Cervical zygapophysial joint pain patterns. I: A study in normal volunteers. Spine 15:453–457.

Grauer JN, Panjabi MM, Cholewicki J, Nibu K, Dvorak J. 1997. Whiplash produces an S-shaped curvature of the neck with hyperextension at lower levels. Spine 22:2489–2494.

Heitplatz F, Sferco R, Fay P, Reim J, Kim A, Prasad P. 2003. 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. Proceedings of the 18th Technical Conference on the Enhanced Safety of Vehicles. Nagoya, Japan. p 1–7.

Kaneoka K, Ono K, Inami S, Hayashi K. 1999. Motion analysis of cervical vertebrae during whiplash loading. Spine 24:763–770.

Klinich KD, Saul RA, Auguste G, Backaitis S, Kleinberger M. 1996. Techniques for developing child dummy protection reference values. National Highway Traffic Safety Administration; Docket Submission #74–14 Notice 97 Item 069.

Lee KE, Thinnes JH, Gokhin DS, Winkelstein BA. 2004. A novel rodent neck pain model of facet-mediated behavioral hypersensitivity: Implications for persistent pain and whiplash injury. J Neurosci Methods 137:151–159.

Macnab I. 1964. Acceleration injuries of the cervical spine. J Bone Joint Surg Am 46:1797–1799.

Miettinen T, Airaksinen O, Lindgren KA, Leino E. 2004. Whiplash injuries in Finland: The possibility of some sociodemographic and psychosocial factors to predict the outcome after one year. Disabil Rehabil 26:1367–1372.

Myklebust JB, Pintar F, Yoganandan N, Cusick JF, Maiman D, Myers TJ, Sances A Jr. 1988. Tensile strength of spinal ligaments. Spine 13:526–531.

Nightingale RW, Carol Chancey V, Ottaviano D, Luck JF, Tran L, Prange M, Myers BS. 2007. Flexion and extension structural properties and strengths for male cervical spine segments. J Biomech 40:535–542.

Ono K, Kaneoka K, Wittek A, Kajzer J. 1997. Cervical injury mechanism based on the analysis of human cervical vertebral motion and head-neck-torso kinematics during low speed rear impacts. Proceedings of the 41st Stapp Car Crash Conf. Lake Buena Vista, FL. p 339–356.

Panjabi M, Wang J-L, Delson N. 1999. Neck injuries criterion based on intervertebral motions and its evaluation using an instrumented neck dummy. Proceedings of the International Research Council on the Biomechanics of Impact (IRCOBI). Sitges, Spain. p 170–190.

Panjabi MM, Cholewicki J, Nibu K, Grauer JN, Babat LB, Dvorak J. 1998. Mechanism of whiplash injury. Clin Biomech 13:239–249.

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

Penning L. 1992. Acceleration injury of the cervical spine by hypertranslation of the head. Part I: Effect of normal translation of the head on cervical spine motion: a radiological study. Eur Spine J 1:7–12.

Pintar FA, Yoganandan N, Voo L. 1998. Effect of age and loading rate on human cervical spine injury threshold. Spine 23:1957–1962.

Pobereskin LH. 2005. Whiplash following rear end collisions: A prospective cohort study. J Neurol Neurosurg Psychiatry 76:1146–1151.

Prasad P, Daniel RP. 1984. A biomechanical analysis of head, neck, and torso injuries to child surrogates due to sudden torso acceleration. Proceedings of the 28th Annual Stapp Car Crash Conf. p 25–40.

Prasad P, Kim A, Weerappuli DPV. 1997. Biofidelity of anthropomorphic test devices for rear impact. Stapp Car Crash Conf. Lake Buena Vista, FL. p 387–415.

Radanov BP, Sturzenegger M, Di Stefano G. 1995. Long-term outcome after whiplash injury. A 2-year follow-up considering features of injury mechanism and somatic, radiologic, and psychosocial findings. Medicine 74:281–297.

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

Spitzer WO, Skovron ML, Salmi LR, Cassidy JD, Duranceau J, Suissa S, Zeiss E. 1995. Scientific monograph of the Quebec task force on whiplash-associated disorders: Redefining "whiplash" and its management. Spine 20:1S–73S.

Stemper BD, Storvik SG. 2010. Incorporation of lower neck shear forces to predict facet joint injury risk in low-speed automotive rear impacts. Traffic Inj Prev 11:300–308.

Stemper BD, Yoganandan N, Pintar FA. 2003. Gender dependent cervical spine segmental kinematics during whiplash. J Biomech 36:1281–1289.

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

Stemper BD, Yoganandan N, Gennarelli TA, Pintar FA. 2005. Localized cervical facet joint kinematics under physiological and whiplash loading. J Neurosurg Spine 3:471–476.

Stemper BD, Yoganandan N, Pintar FA, Maiman DJ, Meyer MA, DeRosia J, Shender BS, Paskoff G. 2008. Anatomical gender differences in cervical vertebrae of size-matched volunteers. Spine 33:E44–E49.

Sterling M, Jull G, Vicenzino B, Kenardy J, Darnell R. 2005. Physical and psychological factors predict outcome following whiplash injury. Pain 114:141–148.

Suissa S. 2003. Risk factors of poor prognosis after whiplash injury. Pain Res Manag 8:69–75.

Sundararajan S, Prasad P, Demetropoulos CK, Tashman S, Begeman PC, Yang KH, King AI. 2004. Effect of head-neck position on cervical facet stretch of post mortem human subjects during low speed rear end impacts. Stapp Car Crash J 48:331–372.

Svensson MY, Aldman B, Hansson HA, Lovsund P, Seeman T, Suneson A, Ortemgren T. 1993. Pressure effects in the spinal canal during whiplash extension motion. Proceedings of the International Research Council on the Biomechanics of Impact (IRCOBI). Eindhoven, Netherlands. p 189–200.

Vasavada AN, Li S, Delp SL. 2001. Three-dimensional isometric strength of neck muscles in humans. Spine 26:1904–1909.

Vasavada AN, Danaraj J, Siegmund GP. 2008. Head and neck anthropometry, vertebral geometry and neck strength in height-matched men and women. J Biomech 41:114–121.

Viano DC, Davidsson J. 2002. Neck displacement of volunteers, BioRID P3 and Hybrid III in rear impacts: Implications to whiplash assessment by a neck displacement criterion (NDC). Traffic Inj Prev 3:105–116.

Yang KH, Begeman PC, Muser M, Niederer P, Walz F. 1997. On the role of cervical facet joints in rear end impact neck injury mechanisms. Warrendale, PA: Society of Automotive Engineers (SAE).

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

Yoganandan N, Pintar FA. 1997. Inertial loading of the human cervical spine. J Biomech Eng 119:237–240.

Yoganandan N, Pintar FA, Klienberger M. 1998. Cervical spine vertebral and facet joint kinematics under whiplash. J Biomech Eng 120:305–307.

Yoganandan N, Pintar FA, Stemper BD, Schlick MS, Philippens M, Wismans J. 2000. Biomechanics of human occupants in simulated rear crashes: Documentation of neck injuries and comparison of injury criteria. Stapp Car Crash J 44:189–204.