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Cervical Facet Capsule Response to Whiplash Loading With a Rotated Head Posture

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ABSTRACT

The cervical facet joints are a source of pain in some chronic whiplash patients. Clinical data show that a head-turned posture increases the severity and duration of whiplash-related symptoms. The objective of this study was to quantify strains in the facet capsule during whiplash-like loading with an axial intervertebral pre-torque simulating an initial head-turned posture. These strains were compared to previously-published strains for partial failure and gross failure of the facet capsule for these specimens to evaluate the potential for capsule injury risk in a whiplash scenario. Thirteen motion segments from seven female donors (50 ± 10 yrs) were exposed to axial pre-torques (±1.5 Nm), axial compressive pre-loads (45, 197 and 325 N) and quasi-static posterior shear loads (0 to 135 N) to simulate whiplash kinematics with the head turned. Three-dimensional displacements of markers placed on the right facet capsular ligament were used to estimate the strain field in the ligament during combined loading. The effects of pre-torque direction, compression and posterior shear on motion segment kinematics and maximum principal strain in the capsule were compared using repeated-measures ANOVAs. Axial pre-torque affected peak maximum principal strains in the capsule more than axial compression or posterior shear. Peak strains reached 34 ± 18% and were higher for pre-torques toward rather than away from the facet capsule. In other words, head rotation to the right produced higher strain in the right facet capsule than rotation to the left. Peak capsule strains were double and similar to those measured at partial failure of the ligament (35 ± 21%) compared to previously-reported data for these specimens under similar shear loads but without a pre-torque. Thus, a head-turned posture increases facet capsular ligament strain compared to a neutral head posture—a finding consistent with the greater symptom severity and duration observed in whiplash patients who have their head turned at impact.

INTRODUCTION

Injury to the cervical facet capsular ligaments is a potential mechanism for chronic pain following whiplash injury, which accounts for half of all patient care expenses from motor vehicle accidents (Quinlan et al., 2004). Distending the facet capsule by injecting contrast media has produced whiplash-like pain patterns in normal individuals (Dwyer et al., 1990), and anesthetic blocks have isolated the cervical facet joints as the source of pain in about half of a chronic whiplash population (Bogduk and Marsland, 1998; Lord et al., 1996). Studies with in-vivo animal models of facet capsule loading have shown that Group III and IV afferents (i.e. pain fibers) from the facet joint capsule exhibit a graded electrical response to mechanical loading of the facet joint in the goat (Lu et al., 2005). Other studies in the rat suggested that a capsular ligament strain-threshold exists above which allodynia—pain in response to a normally non-noxious stimulus—is produced (Lee et al., 2004). These data support a facet capsule-based mechanism for whiplash injury, but do not establish whether human capsular ligaments are injured in the low-speed rear-end collisions to which many whiplash injuries are attributed.

Head rotation is a factor in low-speed rear-end impacts. Whiplash patients who have their head turned at impact have more severe and longer duration symptoms than patients who are facing forward (Sturzenegger et al., 1994, 1995). These findings have prompted follow-up biomechanical studies using human cadaveric necks to investigate why a head-turned posture might increase injury potential. Dynamic rear-impact tests of pre-rotated ligamentous spines (occiput–T1) produce increased neck flexibility (interpreted as injury) in extension, lateral bending and axial rotation (Panjabi et al., 2006). Though concentrated in the lower cervical spine, these "injuries" were not isolated to particular spinal ligaments. Detailed measurements of the strain field in the facet capsule have also shown that a head-turned posture generates higher capsular strains than does a neutral head posture (Winkelstein et al., 2000), but the quasi-static loads applied during those tests were limited to pure flexion/extension moments and did not include the axial compression or posterior shear present during whiplash loading. Thus, the question of how a head-turned posture combined with multi-axial whiplash loads affects facet capsular ligament strain remains unanswered.

Therefore, our goals were to use human cadaveric motion segments to: 1) quantify the intervertebral kinematics and facet capsule strains during vertebral retraction with an axial pre-torque, and 2) compare the capsule strains generated by these combined loads to the previously-published strains for neutral posture and the strains at partial and gross failure. The overall hypothesis of this work was that capsular strains during this simulated whiplash exposure are similar to those needed to injure the capsular ligament.

METHODS

Specimen Preparation

Thirteen motion segments (seven C3/4; six C5/6) from seven unembalmed female human cadavers (50 ± 10 years) were isolated for mechanical testing. The vertebral bodies were cast with the mid-discal plane horizontal using polyester resin and stainless steel wires (Figure 1). Multiple black spherical markers were attached to the specimen for motion analysis: four markers (7.94 mm diameter) to each vertebral body to quantify vertebral motion, two markers (4.76 mm) to the right articular processes immediately superior and inferior to the capsular ligament, and an array of markers (0.79 mm) to the lateral aspect of the right capsular ligament (Figures 1 and 2). Ligament marker arrays varied from 5 x 7 markers to 7 x 7 markers and covered an area of about 1 cm^2 (Figure 2).

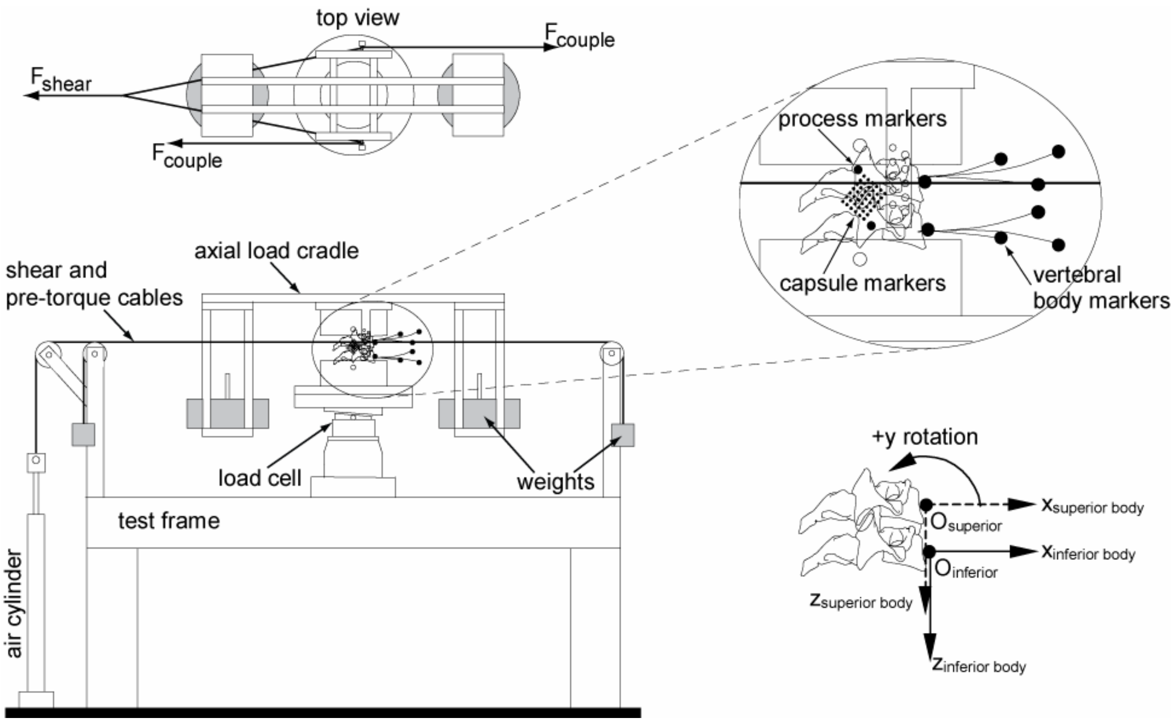

Figure 1: Schematic of the test setup, motion analysis markers and reference frames.

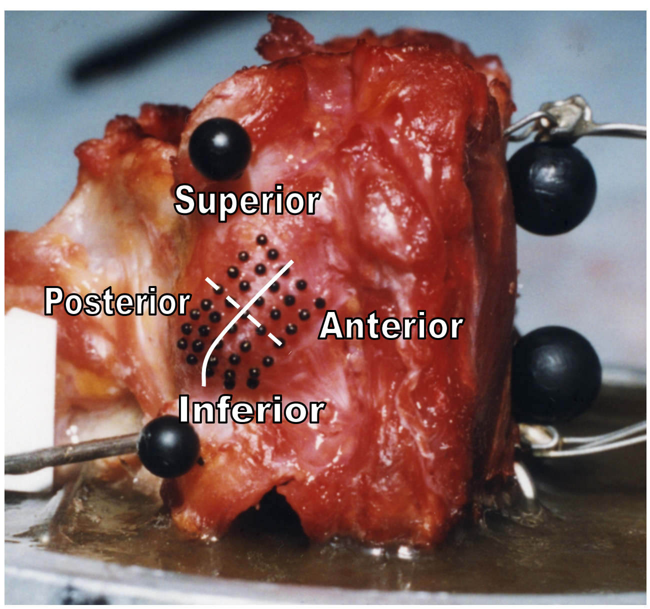

Figure 2: Anatomic quadrants for capsule strain analysis.

Test Equipment

The inferior vertebra was cast in a cup rigidly attached to the test frame through a 6-axis load cell (Denton 1716A, R.A. Denton, Farmington Hills, MI) (Figure 1). Compressive loads were applied statically using weights placed in a cradle attached to the casting cup of the superior vertebra (Siegmund et al., 2001). Axial pre-torques were applied as a force couple to the superior casting cup, and posterior shear loads were applied through a yoke also attached to the superior casting cup (Figure 1) (Winkelstein et al., 2000;

Siegmund et al., 2001). Two 119 x 192 pixel cameras (EktaPro, Eastman Kodak, Charlotte, NC) recorded the vertebral markers and two 640 x 480 pixel cameras (Pulnix TN-9701, Pulnix America Inc, Sunnyvale, CA) recorded the capsule marker arrays.

Test Procedures

Each specimen was pre-conditioned by 30 flexion/extension cycles before undergoing step-wise posterior shear loading (0 to 135 N in 11 steps) under ten different combinations of axial compression (0, 45, 197 and 325 N) and axial pre-torque (0 and ±1.5 Nm) (Table 1). Because the right facet joint was studied in this experiment, a pre-torque to the right (+z) is referred to as an ipsilateral pre-torque and a pre-torque to the left (-z) refers to a contralateral pre-torque. The line of action of the shear force was approximately 8.5 mm above the mid-discal plane, therefore, because of this offset, it also produced an extension moment across the intervertebral joint. Specimens were allowed to creep for 30 seconds at each shear step before load cell and image data were simultaneously acquired. Data from the six pre-load conditions that combine axial compression and axial pre-torque (numbered 5 through 10 in Table 1) are analyzed and reported here; data from the four compression-only pre-load conditions (numbered 1 through 4 in the table) have been reported previously (Siegmund et al., 2001).

Table 1. Test order of the pre-load conditions for all specimens.

	Axial Pre-torque		
Axial Preload	None	+1.5 Nm (ipsilateral)	-1.5 Nm (contralateral)
0 N	1	-	-
45 N	2	5	6
197 N	3	8	7
325 N	4	9	10

The shear loads applied in this study were based on peak horizontal shear forces between 40 and 250 N at the atlanto-occipital joint in human subjects during an 8-km/h rear-end impact (Ono et al., 1997). Compressive loads simulate head weight (45 N; Clauser et al., 1969), a combination of head weight and inertial neck compression developed by torso-seatback interaction in rear-end collisions (197 N; Ono et al., 1997; Siegmund et al., 1997) and the potential combination of head weight, inertial compression and reflexive muscle forces (325 N; Siegmund and Brault, 2001). Axial pre-torques were based on previous work showing that 1.5 Nm produces 2º of rotation (Winkelstein et al., 2000), slightly less than the normal 3 - 4.5º range reported in-vivo at maximal head rotation (Penning and Wilmink, 1987; Ishii et al., 2004).

Data Reduction

Three-dimensional (3D) coordinates of the vertebral and capsule markers were computed from the digitized stereo image pairs using direct linear transformation (Veldpaus et al., 1988). For this study all marker displacements were referenced to the same initial configuration, which corresponded to 45 N of axial compression with no posterior shear or axial pre-torque (condition 2 in Table 1). This reference configuration was selected to mimic a relaxed, forward-facing occupant prior to a rear-end collision. Angular displacements of the vertebra are reported as Cardan-Bryant angles decomposed as flexion/extension first, then axial rotation, and then lateral flexion. Movement of one or more of the four vertebral markers after the 45 N neutral test eliminated four specimens (C3/4 from three donors and C5/6 from one other donor) from the vertebral motion data set. Applied loads are reported relative to the stationary reference frame of the inferior vertebra (Figure 1).

The 3D coordinates of the capsule markers were used to create a mesh of 4-node shell elements. Capsule motion over the curved joint surface obscured some markers and prevented every marker from being used in the mesh. As a result, mesh sizes varied from a grid of 4 x 4 markers to 6 x 6 markers or 5 x 7 markers (Figures 2 and 3). A customized isoparametric mapping program (Matlab 7.2; Mathworks Inc., Natick, MA) was used to compute the deformation gradients, corresponding planar strains and principal strain fields from the 3D displacements of the capsule markers at each applied load. The maximum principal strain (MPS) across all analyzed elements was then determined at each shear step for each load condition.

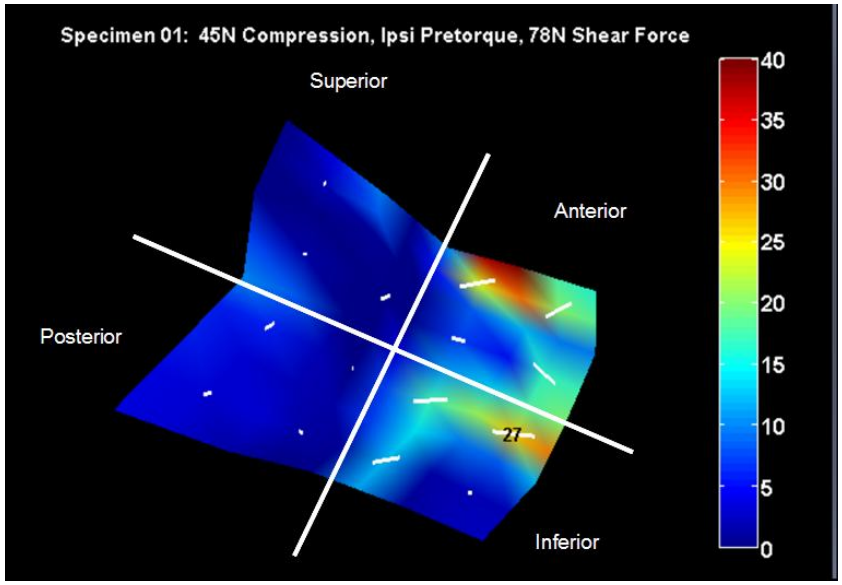

Figure 3: A sample element mesh (5x5 markers) showing principal strain magnitude and direction in the capsule, as well as the anatomic quadrants.

Comparison to Failure Data

For the second part of the analysis of this study, the capsule strains in the combined loading configurations were compared to the strains at minor-rupture and gross-failure reported previously for these specimens (Siegmund et al., 2001). Because some capsule markers were lost during isolation of the facet joint for failure testing and other markers were obscured from one camera's view by intervertebral rotation in the current data set, an element-by-element comparison of strains from the combined-loading and strains at failure was not possible. Therefore, for this analysis each ligament was divided into four quadrants (superior, anterior, inferior, posterior) based on the joint anatomy (Figure 2) and MPS was determined for the elements within each quadrant at each shear step for each combined pre-load condition. For comparison, the previously-published strain data from the failure tests were re-analyzed to define the MPS according to the quadrant regions for this study.

Statistical Analysis

Effects of the combined loading on both intervertebral motion and maximum principal strain in the capsule were analyzed. For intervertebral motion, the effects of axial compression (45, 197, 325 N), posterior shear (0 to 135 N in 11 steps) and axial pre-torque (+1.5, -1.5 Nm) on each of the six degrees of vertebral motion (3 translations: T_x, T_y, T_z; and 3 rotations R_x, R_y, R_z; Figure 1) were tested using a three-way, repeated-measures ANOVA. For MPS in the capsule, the effects of these same three independent variables and capsule quadrant (superior, anterior, inferior, posterior) were tested using a four-way, repeated-measures ANOVA. The distribution of MPS between the four quadrants was compared using pair-wise binomial tests. Comparisons of MPS between the combined-loading condition, and each of partial failure and gross failure were compared using paired t-tests, respectively. All statistical tests were performed using Statistica (v.6.1, Statsoft Inc., Tulsa, OK) and a significance level of $\alpha = 0.05$.

RESULTS

In response to the multi-axial loads applied to the motion segment, the superior vertebra translated in the posterior direction and rotated both rearward (extension) and axially in the direction of the applied pre-torque (Table 2; Figures 4 and 5). In addition to these primary motions, coupled motions in lateral translation and lateral flexion were also observed (Figures 4 and 5). Despite a constant axial pre-torque, the axial rotations established by the initial pre-torques (Figure 5) diminished as the shear load increased and were no longer significantly different at shear loads of 96 and 135 N.

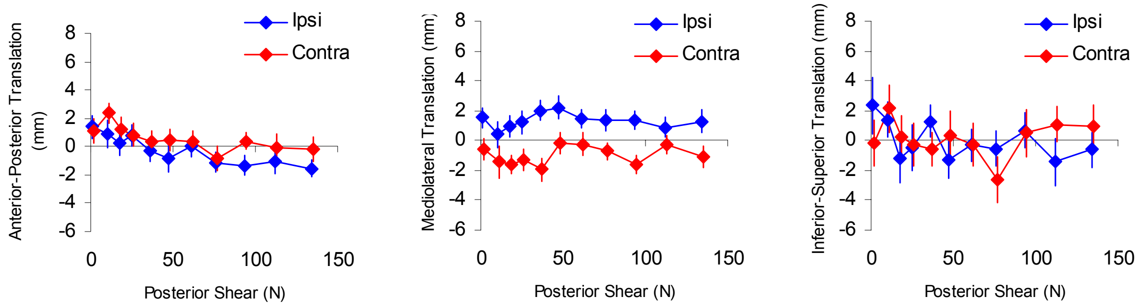

Figure 4: Intervertebral translations. Error bars indicate standard error.

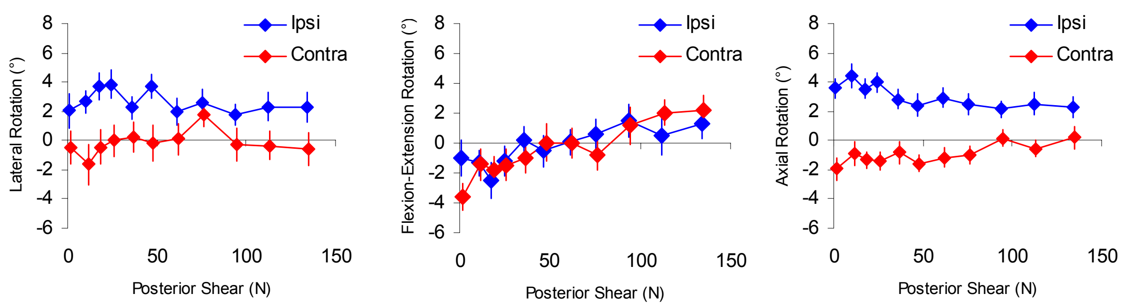

Figure 5: Intervertebral rotations. Error bars indicate standard error.

Table 2. P-values of the significant effects for the 3-way ANOVA used for vertebral motion (T_x through R_z) and for the 4-way ANOVA used for maximum principal strain in the facet capsule (MPS). Interactions not shown were not significant. Interactions are expressed using the first letter of the main effect. The quadrant main effect was not tested for the translation or rotation variables.

Dependent Variable	Statistical analysis (p-values)					
	Main effects				Interactions	
	Quadrant	Compression	Torque	Shear	C × S	T × S
T_x				0.0003		
T_y			0.043			
T_z						
R_x			0.025			
R_y		0.0005		<0.0001		
R_z			0.0002			0.002
MPS	0.023		0.020		<0.0001	0.0005

Maximum principal strain did not vary with compressive preload, Table 2. MPS in the facet capsule was affected more by axial pre-torque than by either axial compression or posterior shear (Table 2; Figure 6A). At all but the lowest shear loads, MPS was higher for the ipsilateral pre-torque than for the contralateral pre-torque (Table 2, Figure 6a) and reached a maximum of 34 ± 18% at 96 N of shear when pooled across compressive preloads. Within-specimen differences between the ipsilateral and contralateral strains increased with shear at low shear loads (Figure 6b), but like axial rotation, these diminished and were not significantly different at the higher shear loads.

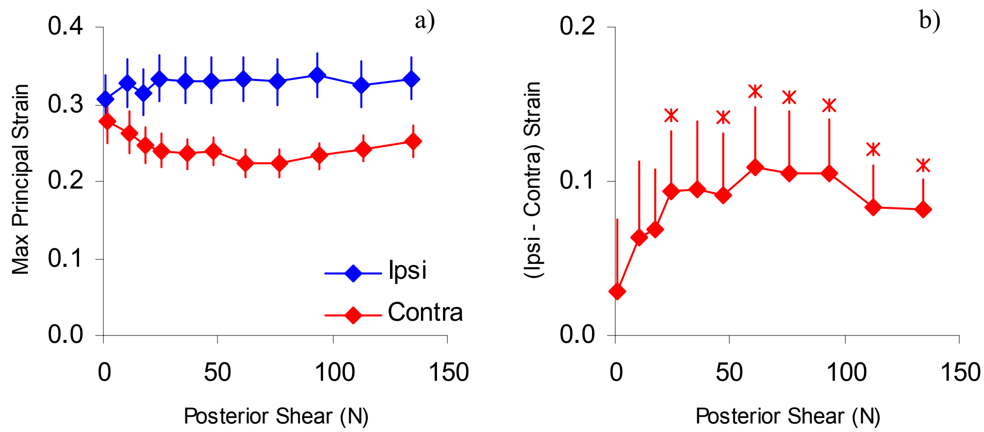

Figure 6: (a): Average maximum principal strain in the capsule, grouped by pre-torque. (b): Difference between average maximum principal strain for the ipsi and contra pre-torques. Asterisks indicate values that are significantly different from zero.

Maximum principal strain in the facet capsule was most frequently observed in elements located in the superior and anterior quadrants of the capsule. Across all shear steps, the magnitude of the MPS was less in the inferior quadrant of the capsule than in either the anterior or superior quadrants (post-hoc, p=0.045 and 0.054, respectively; Figure 7). For the ipsilateral pre-torque, the location of MPS at zero shear occurred most frequently in the superior and anterior quadrants (Figure 8a), although a similar pattern was not observed for the contralateral pre-torque (Figure 8c). At maximal posterior shear, the location of the MPS occurred most frequently in the superior quadrant for the ipsilateral condition and in the anterior quadrant for the contralateral condition (Figure 8b,d). MPS at maximum posterior shear occurred least frequently in the inferior quadrant under both pre-torque conditions (Figure 8).

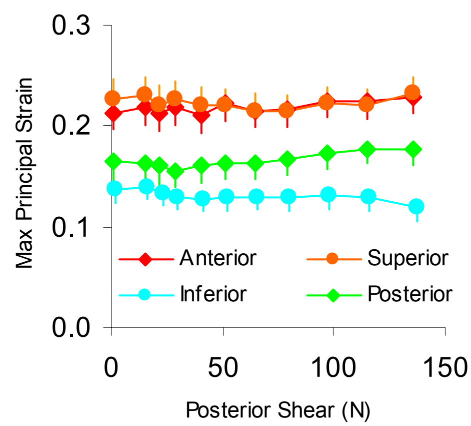

Figure 7: Average maximum principal strain in the anatomic quadrants.

When the previously-published failure data were partitioned into the four quadrants defined here, MPS at gross failure (previously reported to be 0.94 ± 0.85; Siegmund et al., 2001) occurred most frequently in the superior and inferior quadrants (Figure 9). MPS at partial failure (previously reported to be 0.35 ± 0.21) occurred most frequently in the inferior quadrant (Figure 9a). A quadrant-by-quadrant comparison revealed that peak strains produced by the multi-axial whiplash-like conditions used here reached 84±51% of the strains causing partial failure ($p=0.19$) and 66±83% of the strains causing gross failure ($p=0.02$). A within-specimen comparison of the MPS between the multi-axial loading tests and failure tests indicated that two specimens exceeded their partial failure strains and two other specimens exceeded their gross failure strains during the whiplash-like loads used here.

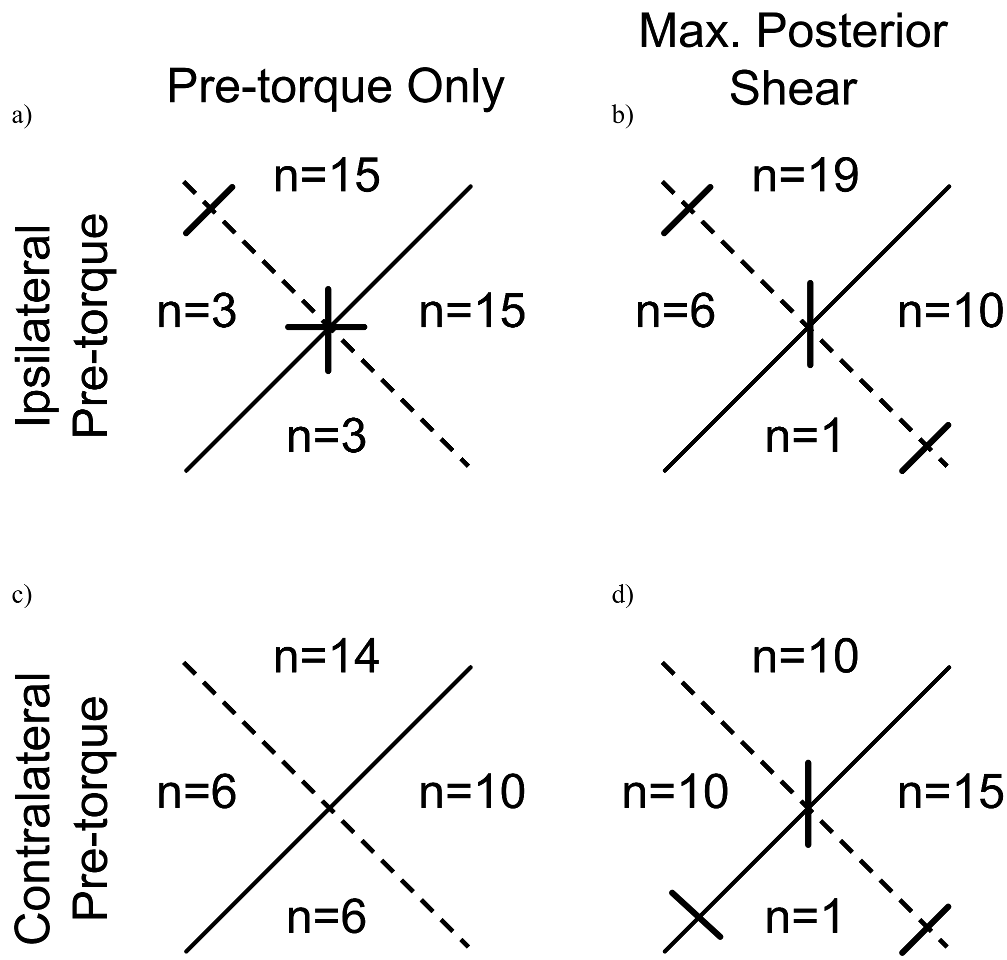

Figure 8: Distribution of the maximum principal strains within the four quadrants of the capsular ligament. Data shown are from the whiplash-like loading tests (top two rows) for the ipsilateral (a,b) and contralateral (c,d) pre-torque conditions with no posterior shear (a,c) and maximum posterior shear (b,d). Solid and dashed lines depict the capsule quadrants. Lines between quadrants show significantly different proportions based on binomial comparisons ($p<0.05$).

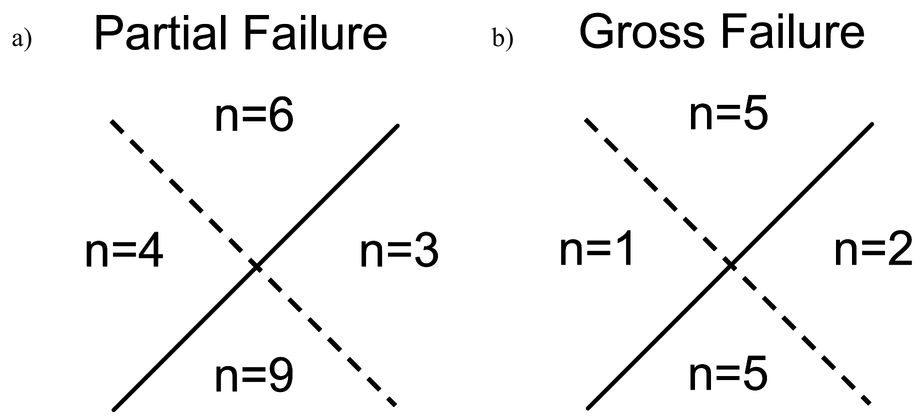

Figure 9: Distribution of the maximum principal strains within the four quadrants at partial and gross failure. Data from the failure tests include 22 partial failures (a) observed before gross failure (b) of the 13 ligaments. Solid and dashed lines depict the capsule quadrants.

DISCUSSION AND CONCLUSIONS

Axial pre-torque and the resulting axial rotation of the intervertebral joint have a large effect on the maximum principal strain in the cervical facet joint capsule when combined with compression, shear and extension loads simulating a low-speed rear-end automobile impact. Peak strains in the capsule with an ipsilateral pre-torque (34 ± 18%) were double the previously-reported peak strains without a pre-torque (17 ± 6%) and not significantly different from the previously-reported strains to cause partial failures (35 ± 21 %) in these specimens (Figure 10) (Siegmund et al., 2001). These findings are consistent with the increased severity and duration of whiplash symptoms in patients who had their head turned at the time of impact (Sturzenegger et al., 1994, 1995) and provide a possible explanation for these clinical findings.

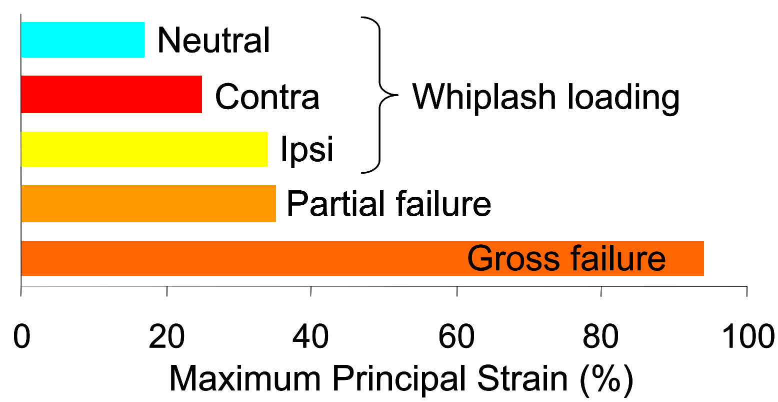

Figure 10: Comparison of maximum principal strains for the different test conditions

With the addition of posterior shear to the axial pre-torque, maximum principal strain increased by about 3% in the ipsilateral condition, but initially decreased by about 6% in the contralateral condition before beginning to increase (Figure 6a). Maximum principal strain in the capsule also occurred most frequently in the superior and anterior quadrants of the capsule with the ipsilateral pre-torque and then concentrated in the superior quadrant when shear was added (Figures 8a,b). In contrast, maximum principal strain did not occur more frequently in any particular quadrant with the contralateral pre-torque, but then did concentrate in the anterior quadrant when shear was added (Figures 8c,d). This combination of findings suggests that posterior shear and the ipsilateral pre-torque strain similar parts of the capsule and thus maximum principal strain in the capsule continues to increase when shear is added. Conversely, posterior shear and the contralateral pre-torque appear to strain different parts of the capsule and thus maximum principal strain in the capsule is initially relieved when shear is added (Figure 6a). Viewed more broadly, these findings suggest that the facet capsules located on the side of the neck towards which a vehicle occupant's head is turned are most likely to be injured in a rear-end crash, although we could find no clinical or epidemiological data to support or refute this specific proposition.

Increased axial compression of the cervical spine has been shown to increase flexibility to horizontal shear (Yang et al., 1997), and this increased flexibility has been postulated to increase neck injury potential. In the current study, we did not observe increased flexibility in posterior shear with increased axial compression; on the contrary, we observed lower extension angles at the 197 and 325N compression levels. The reason for these differences between studies is not clear, but may be related to the use of full cervical spines by Yang et al. (1997) and the use of motion segments here. Capsular strains were not affected by axial compression, and were greater in the presence of both axial pre-torques than in the neutral position for all levels of axial compression.

Intra-specimen comparisons indicate that during the whiplash-like loads applied here 2 of 13 specimens (15%) exceeded the strains needed to cause partial failure of the capsule. The potential for 15% of specimens to exceed a partial failure threshold is consistent with earlier quasi-static work (Winkelstein et al., 2000; Siegmund et al., 2001) and more recent dynamic work (Pearson et al., 2004). Similar levels of capsule strains have also produced behavioral and electrophysiological evidence of short and long term pain in animals (Lee et al., 2004; Lu et al., 2005), although both animal experiments strained the dorsal aspect of the capsule rather than the lateral aspect studied here. The 15% risk of partial failure in the capsule indicated by the current data is similar to the 12% risk of whiplash-exposed individuals suffering chronic symptoms (>6 months; Suissa et al., 2001), though considerable work remains to determine whether these similar risk values are related or coincidental.

Two other specimens experienced maximum principal strains during the combined whiplash-like loads that exceeded their maximum principal strain at gross failure. However, there was no evidence of gross failure during the whiplash-like loading, and thus this finding likely highlights limitations in our technique. We previously assumed that partial failures and gross failures occurred in the element with the highest maximum principal strain (Siegmund et al., 2001). Yet here we compare whiplash and failure strains quadrant-by-quadrant rather than element-by-element because of lost and/or replaced markers on the capsule. Since the capsular ligament is not a homogeneous structure, this assumption is a limitation of our work and may be incorrect. In addition, regional differences in the ligament could result in different mechanical tolerances at different locations within a quadrant or even within an element (Quinn and Winkelstein, 2007). Moreover, the failure tests were conducted along the antero-posterior axis of the facet joint, whereas the whiplash tests exposed the facet joint to compound 3D displacements. This means that different fibers within the ligament may have borne the loads during the whiplash and failure tests. In fact, the quadrant where the MPS was most often located differed between the whiplash-like loading and the failure loading. Thus even though our experimental technique provides more detailed strain-field information than other recently-published techniques (Pearson et al., 2004; Panjabi et al., 2006), even finer techniques—perhaps looking at region-specific or even fiber-specific strains—are likely needed to capture regional differences and properly characterize capsular ligament behavior during whiplash (Quinn and Winkelstein, 2007).

In summary, we examined the intervertebral kinematics and facet capsule strains under whiplash-like loading conditions in the presence of an initial axial intervertebral rotation. We found that an axial rotation of the head doubles the maximum principal strain in the capsular ligament compared to the neutral posture. We also found that capsular strains during the simulated whiplash exposure with the head turned are not significantly different from maximum principal strains associated with partial failure of the capsule. Thus these findings support the overall hypothesis that excessive capsular strains experienced by some individuals during some whiplash conditions may be responsible for painful capsular whiplash injury.

ACKNOWLEDGEMENTS

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DISCUSSION

PAPER: **Cervical Facet Capsule Response to Whiplash Loading With a Rotated Head Posture**

PRESENTER: ***Martin Davis, Department of Neurosurgery, University of Pennsylvania***

QUESTION: *Guy Nusholtz, DaimlerChrysler*
You mentioned that it may not represent the dynamic situation. Could you pontificate on how you think it could or what would be the issues with trying to reproduce dynamic? Obviously, in one case, you have a much shorter duration type of pulse; and in addition, you're going to have a time-varying load instead of a static type of load.

ANSWER: Sure.

Q: Do you think there's relevance to this with regard to a dynamic situation?

A: I think dynamically you'd see a difference in that what we saw with the, sort of, unwinding, that pre-torque was kind of taken away by the quasi-static sheer. I'm not so sure you'd see that effect in a dynamic case. So, I think the head inertia would be a little different and you might get—you might not see that same type of unwinding.

Q: And do you expect to see the same type of strains that you would see in a dynamic? I would think they'd be different.

A: I think we know the difference between quasi-static and dynamic testing. Well, at least if you look at failure, you don't see, necessarily, the differences in strains going up to failure. I'm not so sure. I mean that's a good question. Let's try to answer that.

Q: Okay. Thank you.

Q: *Costin Untaroiu, University of Virginia*
I didn't understand how you calculated the principal strain. You said something. You use finite element or you use at points in model?

A: Yes. So we calculated the displacement of those markers at each step and compared that back to our reference frame. So it's all displacement based strain calculations.

Q: So you measure, on exterior, right?

A: Yes. So we put the markers on the exterior surface of the capsule. They were glued on there. And then based on where they moved, we calculated the displacement differences from the initial configuration.

Q: Did you try with a real finite element model to see if its actually maximum strain?

A: No, we haven't done that sort of modeling for this. No. This is all experimental.

Q: *Frank Pintar, Medical College of Wisconsin*
How much head rotation would you need to be able to get the kind of pre-torques that you're talking about? I mean, obviously most of your head rotation early on occurs from between C1 and C2.

A: Sure.

Q: So how—do you have to turn your head all the way backwards to--?

A: No. These—That one and a half Newton meter pre-torque put the—For those vertebral levels C3/4 and C5/6, put it within physiological range of motion. I think it was on the order of 2 to 4°. Maybe something like that. So that was an appropriate pre-load for those levels for physiologic range - we're just really turning it out of the way.

Q: Two to 4° for that segment. Right?

A: Yes. For that cervical level, yes.

Q: But you don't know how that would translate into how much the head would actually have to rotate to get that 2-4° at that segment?

A: Oh, if you look global like that? I'm not sure. I think those values—We wanted to put on something that was physiologic at those levels, and so that magnitude of load was chosen so that we were within, you know, normal range. I couldn't speak to that global type of motion.

Q: Okay. Thanks.