Timothy P. Holsgrove

Department of Bioengineering, School of Engineering and Applied Science, University of Pennsylvania, 210 South 33rd Street, Room 240 Skirkanich Hall, Philadelphia, PA 19104 e-mail: thols@seas.upenn.edu

Nicolas V. Jaumard

Department of Neurosurgery, Pennsylvania Hospital, University of Pennsylvania, Washington Square West Building, 235 South 8th Street, Philadelphia, PA 19106 e-mail: njaumard@gmail.com

Nina Zhu

Department of Bioengineering, School of Engineering and Applied Science, University of Pennsylvania, 210 South 33rd Street, Room 240 Skirkanich Hall, Philadelphia, PA 19104 e-mail: nzhu@seas.upenn.edu

Nicholas S. Stiansen

Department of Bioengineering, School of Engineering and Applied Science, University of Pennsylvania, 210 South 33rd Street, Room 240 Skirkanich Hall, Philadelphia, PA 19104 e-mail: nsti@seas.upenn.edu

William C. Welch

Department of Neurosurgery, Pennsylvania Hospital, University of Pennsylvania, Washington Square West Building, 235 South 8th Street, Philadelphia, PA 19106 e-mail: william.welch@uphs.upenn.edu

Beth A. Winkelstein¹

Department of Bioengineering, School of Engineering and Applied Science, University of Pennsylvania, 210 South 33rd Street, Room 240 Skirkanich Hall, Philadelphia, PA 19104; Department of Neurosurgery, Pennsylvania Hospital, University of Pennsylvania, Washington Square West Building, 235 South 8th Street, Philadelphia, PA 19106 e-mail: winkelst@seas.upenn.edu

Upper Cervical Spine Loading Simulating a Dynamic Low-Speed Collision Significantly Increases the Risk of Pain Compared to Quasi-Static Loading With Equivalent Neck Kinematics

Dynamic cervical spine loading can produce facet capsule injury. Despite a large proportion of neck pain being attributable to the C2/C3 facet capsule, potential mechanisms are not understood. This study replicated low-speed frontal and rear-end traffic collisions in occiput-C3 human cadaveric cervical spine specimens and used kinematic and fullfield strain analyses to assess injury. Specimens were loaded quasi-statically in flexion and extension before and after dynamic rotation of C3 at 100 deg/s. Global kinematics in the sagittal plane were tracked at 1 kHz, and C2/C3 facet capsule full-field strains were measured. Dynamic loading did not alter the kinematics from those during quasi-static (QS) loading, but maximum principal strain (MPS) and shear strain (SS) were significantly higher (p = 0.028) in dynamic flexion than for the same quasi-static conditions. The full-field strain analysis demonstrated that capsule strain was inhomogeneous, and that the peak MPS generally occurred in the anterior aspect and along the line of the C2/ C3 facet joint. The strain magnitude in dynamic flexion continued to rise after the rotation of C3 had stopped, with a peak MPS of $12.52 \pm 4.59\%$ and a maximum SS of $5.34 \pm 1.60\%$. The peak MPS in loading representative of rear-end collisions approached magnitudes previously shown to induce pain in vivo, whereas strain analysis using linear approaches across the facet joint was lower and may underestimate injury risk compared to full-field analysis. The time at which peak MPS occurred suggests that the deceleration following a collision is critical in relation to the production of injurious strains within the facet capsule. [DOI: 10.1115/1.4034707]

Keywords: cervical spine, facet, injury, dynamic, strain

¹Corresponding author.

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1 Introduction

Dynamic loading of the upper cervical spine can lead to injury, instability, and pain [1–3]. Neck injury is the most common injury in motor vehicle occupants requiring hospital emergency department treatment in the U.S. [4]. It has been estimated that between one-quarter and one-third of people exposed to rear-end collisions, which commonly occur at speeds of 15-30 km/h, experience neck injury [5], and that 15-40% of those injuries will result in chronic neck pain, which can extend to the head, shoulders, and arms [6–8]. Although chronic neck pain is prevalent among the general population [9,10], exposures like those due to low-speed collisions have been reported to increase the incidence by 2.7 times [11].

The origin of a variety of pain symptoms can be traced to specific regions of the spine [9,12,13]. Nerve blocks and/or injections and provocative studies demonstrate that for 35–60% of people with chronic neck pain, the source of the pain is the cervical facet joint [3,10,12,14]. Moreover, the C2/C3 and C5/C6 cervical spinal levels are those most commonly identified as painful in both idiopathic neck pain [12] and in chronic neck pain from whiplash exposure [3,14].

Any disruption to the various hard and soft tissue structures of the facet joint has the capacity to elicit pain [2,15,16]. The facet capsule and synovial folds are innervated by nociceptive and mechanoreceptive afferents [17–21]. Pain can result from direct damage of nociceptors [22–24], but can also be produced indirectly through damage to the mechanoreceptors, which alters feedback and increases neck instability, leading to pain in muscles and/or from muscular contractions [1,22,25–27].

Previous studies of facet injury from dynamic neck loading, such as whiplash exposure, have focused on defining the global and local kinematics of the cervical spine [28-30], deformation of the facet capsule [25,31-33], and acute and chronic pain responses [24,34,35]. Human volunteer studies simulating the rear-end collisions define a characteristic response, with the torso moving prior to the head, leading the neck to undergo an S-shaped configuration approximately 60–100 ms after vehicle impact, followed by the rearward rotation of the head extending the neck approximately 85-140 ms after impact, before the head and torso rebound due to deceleration following the impact, combined with the effect of seat support and muscle activation [30,36]. Simulations of rearend collisions using human cadaver models demonstrate similar kinematic patterns over the same time period [28,37-39] and also show that the facet capsule linear strain increases at C3/C4 and

C6/C7 with an impact acceleration of 8 g compared to a physiological rate of spine loading [25]. More detailed strain measurements of the lower human cervical spine have been estimated using stereophotogrammetry to define the three-dimensional (3D) full strain-field of the facet capsule and demonstrate that physiological loading of the spine can induce strains sufficient to induce injury [32,40]. A key finding of both the human volunteer and cadaveric studies is that intervertebral rotations do not exceed the normal physiological range and that injuries result from *abnormal* kinematics, such as altered intervertebral rotation axes, which can increase the loading in and across the facet joint [25,36].

The subcatastrophic and catastrophic failure of human facet capsules is highly variable [32,40-43]. Despite this, subcatastrophic strains of 35-67% are associated with injury and/or injury potential [32,40,41], which is within the range measured during in vitro studies of rear-end collisions in the lower cervical spine [25]. But these studies do not account for the large proportion of exposures in which pain is attributed to the upper cervical spine, and specifically the C2/C3 level [3,14]. In vivo studies of the neural activity in the facet capsule of goats indicate that the facet capsule has both low- and high-strain threshold units, which discharge at 10% and 47% strain, respectively [44]. Some lowthreshold units are likely to play a role in proprioception, but capsular strains as low as 19% have been shown to induce pain in the absence of any failure of the facet capsule [35] when imposed at rates (500%/s) comparable to those sustained during whiplash exposures [45,46]. As such, defining facet strains in the context of the potential for pain and relative to typical mechanical metrics of tissue failure is necessary to understand injury risk in humans.

Despite the prevalence of neck injury resulting from dynamic loading events, and the large proportion of such injuries being reported in the upper cervical spine [3,14], most investigations of the facet capsule response have been focused in the lower cervical spine. The full strain-field of the C2/C3 facet capsule during any loading exposure, regardless of rate, has not yet been defined nor have the spinal kinematics been used to contextualize facet behavior during potentially injurious exposures with respect to equivalent vertebral rotations during normal physiological loading. Accordingly, this study uses a human cadaver model to investigate the response of the C2/C3 facet to dynamic flexion–extension loading of the upper cervical spine and compares the vertebral kinematics and full-field strains at C2/C3 with those responses under quasi-static loading.



Fig. 1 Global view (*a*) of the occiput (Occ)-to-C3 specimen, with tracking markers at each level. The Occ was fixed to a phantom head. The C3 vertebra was rigidly fixed to the cradle, which was actuated for dynamic tests, with all other levels unconstrained in the sagittal and axial planes. The C2/C3 facet capsule was imaged (*b*) with two cameras, from which 2D facet kinematics were measured (*c*), and 3D reconstructions were used to measure the maximum principal strain (MPS) (*d*) and shear strain.

^{121006-2 /} Vol. 138, DECEMBER 2016

2 Materials and Methods

2.1 Specimen Preparation. Fresh frozen occiput-C3 specimens (n = 6; three males and three females), 66 ± 7 years of age, were used in this study, having been acquired by appropriate methods and procedures for our institution. Prior to testing, the specimens were thawed overnight at room temperature while in sealed bags; all the musculature was dissected and only the disks, ligaments, and facet capsules were left intact. Self-tapping screws were driven into the occiput, and self-tapping screws combined with Kirschner wires were driven into the vertebral body and spinous process of C3 to fix the specimen into pots along with a twopart fast-curing liquid polymer (Smooth Cast® 300, Smooth-On, Inc., Macungie, PA). The C3 vertebra was potted with the center of the vertebral body aligned with the superior end of the specimen pot, and with an anterior rotation of 25 deg, based on the natural angular orientation of that vertebra in the normal lordosis with the head in the neutral position [47]. A surrogate head with a mass (4.15 kg) and moment of inertia in the sagittal plane (0.0210 kg m^2) representative of an average human head [48] was fixed to the occiput specimen potting fixture to account for the inertial effects of the head mass during loading. The center of gravity of the surrogate head was located approximately 60 mm above, and 20 mm anterior to, the occipital condyles [48].

2.2 Test System. Tests were performed using a customdeveloped cradle assembly actuated via a servo-hydraulic testing machine (370.02 FlexTest 60, MTS Systems Corp., Eden Prairie, MN) (Fig. 1). A linkage between the servo-hydraulic actuator and the edge of the cradle assembly converted linear motion in the actuator into angular motion in the cradle assembly. The center of rotation of the cradle assembly was located in a fixed position approximately 16 mm below the center of the C3 vertebral body for each specimen when in the neutral position; this orientation introduced a combination of rotation and translation to the C3 vertebral during dynamic tests. The actuator of the testing machine was fitted with a 5g capacity uniaxial accelerometer (Model 7521A2, Dytran Instruments, Inc., Chatsworth, CA) to measure the acceleration and deceleration during dynamic tests. Four markers were attached to the surrogate head and three additional markers were attached to the loading cradle to track their motions in order to obtain the rotation of the C3 vertebra, and the global range of motion (ROM) of the specimen. Distracting pins (12 mm threaded head) with four reflective polypropylene markers were screwed into the C1 and C2 in an anterior-posterior orientation (Fig. 1(a)), which were used for motion tracking. The C2/C3 facet joint-line was identified via visual inspection and gentle manipulation of the joint, and 1 mm opaque-black painted steel beads were fixed in an array over the outer surface of the entire capsule using a minimal amount of cyanoacrylate glue (Fig. 1(b)). Positioning of the marker array with respect to the joint-line was confirmed using lateral radiographs of each specimen. The C3 was rigidly fixed to the loading cradle (Fig. 1). Pins attached to the surrogate head fitted between two acrylic guide rails, which constrained the specimen in the coronal plane, but allowed free movement in the sagittal and axial planes.

One high-speed camera (V4, Vision Research, Inc., Wayne, NJ), with 256×256 pixel resolution, tracked the two-dimensional (2D) global kinematics of the specimen in the sagittal plane, which enabled measuring the vertebral marker displacement and the intervertebral rotation at each level. The images were calibrated against the known distance between the markers on the actuation cradle, providing a scale of 0.87 ± 0.02 pixels/mm. High-speed stereophotogrammetry was adopted using two cameras (Miro eX1; 480×360 pixel resolution, Vision Research, Inc.), which tracked the marker array on the C2/C3 facet capsule to enable the 3D reconstruction of facet capsule deformation (Fig. 1(*d*)). The dual-camera array was calibrated prior to testing each specimen using a standard marker array and had an error of 0.16 ± 0.12 mm. All the cameras were synchronized; the position

and acceleration of the actuator on the MTS and all the image data were acquired at 1 kHz for all the tests.

2.3 Test Protocol. Each specimen underwent quasi-static and dynamic testing in flexion and extension: quasi-static testing replicated normal physiological flexion–extension, and the dynamic tests simulated low-speed front- and rear-end collisions. Quasi-static tests were completed before and after dynamic tests to assess if there were any effects of the dynamic loading on vertebral kinematics. A 5-min recovery period was allowed between each test, with the specimen fixed in the neutral position.

The quasi-static tests were performed by commencing data acquisition prior to manually guiding the surrogate head through flexion and extension with C3 remaining stationary. The global ROM applied to specimens was 16-20 deg in flexion and 13-20 deg in extension. Dynamic loading consisted of the surrogate head being held in the neutral position by an electromagnet, which was released simultaneously with the actuation of the C3 vertebra via the loading cradle. The loading cradle was used to apply 6.1 deg of rotation to C3 at 100 deg/s and approximately 1.7 mm of translation, which replicated the rotation of the C3 vertebra during a low-speed collision [49]. The actuation was applied using a square wave, with the accelerometer on the actuator used to determine the acceleration and deceleration to/from the test rate equating to 100 deg/s. A separate dynamic test was performed to actuate C3 in flexion combined with anterior translation, and extension combined with posterior translation, which replicated rear- and front-end collisions, respectively. A trigger was used to simultaneously release the electromagnet and actuate the loading cradle, as well as to initiate data acquisition.

2.4 Data Analysis. The displacements of the markers defining the vertebral kinematics and the facet capsule deformation were analyzed for all the tests using ProAnalyst software (Version 1.5.7.7, Xcitex, Inc., Woburn, MA). The 3D displacement data of the markers on the facet capsule were used to estimate the full-field Lagrangian strain using LS-DYNA software (Version R7.0.0, Livermore Software Corporation, Livermore, CA), with the position of the markers 20 ms prior to actuation serving as the 0% strain reference condition.

The two sets of quasi-static flexion–extension tests for each specimen were compared at each vertebral level using 1 deg increments of global ROM from 16 deg of flexion to 13 deg of extension. Comparisons of the rotation at each cervical level (Occ/C1, C1/C2, and C2/C3) relative to the global ROM were made using two-way repeated measures ANOVAs (IBM SPSS Statistics 22.0.0.1, IBM Corporation, Armonk, NY) with a significance level of 0.05. The 3D full-field strains during dynamic loading were assessed over a 150 ms window, with the actuation occurring at 20–81 ms. The maximum principal strain (MPS) and shear strain (SS) were calculated for each element of the facet capsule array at 10 ms intervals. In order to include the strain behavior at the time that the actuation was stopped, measurements at 81 ms after the test started (61 ms after C3 actuation began) were used in preference to the 80 ms time interval.

All the comparisons of the MPS and SS between dynamic and quasi-static loading were performed using Wilcoxon signed-rank tests (IBM SPSS Statistics) with significance at 0.05. For each specimen, the peak MPS during dynamic loading was identified and compared with the peak MPS during quasi-static loading at equivalent C2/C3 rotation. In addition, for each specimen, the mean MPS across all the elements of the facet capsule at the time of peak MPS was calculated and compared with the mean MPS during quasi-static loading to assess whether the peak capsule strains were similar to the overall facet capsule behavior. In addition to statistical comparisons of MPS magnitude, the direction and location of the peak MPS on the facet capsule were identified to assess the strain characteristics between specimens, and loading conditions. The maximum and minimum SS during dynamic

Journal of Biomechanical Engineering

flexion and extension were also compared with the SS during quasi-static loading at equivalent C2/C3 rotations. To account for the curvature of the facet capsule, the SS was estimated in all the three anatomical planes, and the peak value was taken as the maximum. Both the maximum and minimum SS were measured to account for positive and negative shear occurring at different periods within the exposures. Comparisons were made between the maximum and minimum SS during dynamic testing with equivalent quasi-static data using Wilcoxon signed-rank tests (IBM SPSS Statistics) and significance at 0.05.

In order to relate the facet behavior under dynamic loading to previously published in vitro cadaveric studies, the position of the most superior and inferior rows of markers was used to calculate the 2D facet joint posterior-anterior sliding, compressionseparation, and linear strain across the C2/C3 facet using methods previously described [25] (Figs. 1(b) and 1(c)). Posterior sliding was taken as positive, facet separation was defined as the maximum distance across the facet, compression defined as the minimum distance across the facet, and the linear facet strain was defined as the distance between corresponding pairs of markers on the superior and inferior facets relative to the distance 20 ms prior to actuation of C3. The maximum posterior-anterior sliding, separation-compression, and linear strain across the facet each were calculated at the same time increments as the full-field strain analysis, and statistical comparisons were made with equivalent quasi-static data using Wilcoxon signed-rank tests (IBM SPSS Statistics) with significance at 0.05.

3 Results

Dynamic flexion-extension did not significantly alter the intervertebral rotations compared to quasi-static loading at any cervical level (p=0.997 for Occ/C1; p=0.999 for C1/C2; andp = 0.714 for C2/C3) (Fig. 2). The average acceleration/deceleration of the MTS actuator to/from the test rate equating to 100 deg/ s during dynamic tests was 2.8 \pm 0.8 g and 2.8 \pm 0.8 g in dynamic flexion and 1.8 ± 0.2 g acceleration and 2.8 ± 0.3 g deceleration in dynamic extension. All the specimens exhibited similar kinematic patterns at each level during dynamic loading in flexion and extension, although the magnitude of rotation at each level did vary between specimens (Fig. 3). Dynamic loading altered the rotation at the C2/C3 level from extension to flexion in the flexion tests and from flexion to extension in the extension tests (Fig. 3). This change occurred shortly after the actuation stopped at 81 ms. After 150 ms during the dynamic tests, all the specimens moved into flexion due to the moment resulting from the mass and the position of the center of gravity of the surrogate head. The maximum and minimum rotation at each level was generally less than the rotation during quasi-static flexion-extension tests (Table 1) and was within physiological ROMs previously reported [50,51].

In dynamic flexion, the peak MPS in the capsule was $12.52 \pm 4.59\%$ and the mean MPS was $6.59 \pm 2.93\%$. In dynamic

extension, these strains were $7.02 \pm 1.88\%$ and $2.08 \pm 0.89\%$, respectively. The peak MPS occurred at the time when dynamic actuation stopped (81 ms) or later in all the flexion tests, but was more varied when extension was applied, occurring both before and after actuation was stopped (range 40–150 ms) (Fig. 4). The peak MPS was significantly higher during dynamic flexion than during quasi-static loading (p = 0.028), whereas there was no difference between peak MPS in dynamic extension and quasi-static loading (p = 0.600) (Fig. 5(*a*)). The same statistical outcomes were found between the mean MPS during dynamic and quasi-static loading. The mean MPS was significantly higher in dynamic flexion than during quasi-static loading (p = 0.028), but there was no difference between dynamic extension and quasi-static loading (p = 0.600) (Fig. 5(*b*)).

The location of the peak MPS was in the center and toward the anterior aspect of the facet capsule in dynamic flexion (Fig. 6). In dynamic extension, it was largely located in the center and toward the posterior aspect of the capsule. The direction of the peak MPS was orientated approximately along the facet joint, particularly in the region of capsule where the maximum strain was sustained. The general position and direction of the MPS were similar to dynamic conditions at equivalent quasi-static rotations, but the magnitude was less pronounced relative to the surrounding elements (Fig. 6).

The greatest shear strain generally occurred in the sagittal plane in all the specimens throughout dynamic loading, although those with more pronounced curvature of the facet exhibited maximum/ minimum SS in the coronal and axial planes at some time increments. In dynamic flexion, the maximum SS was $5.18 \pm 1.48\%$ and the minimum was $-4.30 \pm 2.73\%$. In dynamic extension, these strains were $2.65 \pm 1.65\%$ and $-4.38 \pm 1.50\%$, respectively. The greatest SS changed from a minimum to a maximum in dynamic flexion, with the reverse occurring in dynamic extension; these changes corresponded approximately to the time when the actuation of C3 stopped (at 81 ms) (Fig. 4). One specimen was omitted from the comparison of the maximum SS in dynamic flexion, and a separate specimen was omitted from the dynamic extension comparison due to no C2/C3 rotations during quasi-static loading being within 0.1 deg of the C2/C3 rotation during dynamic loading. The maximum SS was significantly greater in dynamic flexion (p = 0.028) and extension (p = 0.028) compared to quasi-static loading, but there was no difference in the minimum SS (Table 2).

The facet sliding was consistent across all the pairs of markers from the anterior to posterior aspect of the facet, so the magnitude was averaged at each time increment for each specimen (posterior sliding taken as positive). The separation, compression, and linear strain varied across the facet; therefore, the greatest value at each time point was used for analysis. Posterior–anterior facet sliding exhibited similar behavior to the rotational kinematics, with the facet undergoing posterior sliding during actuation in dynamic flexion and changing to anterior sliding at approximately the time



Fig. 2 Mean (±95% confidence intervals (CI)) quasi-static kinematics (flexion positive) at the Occ/C1 (*a*), C1/C2 (*b*), and C2/C3 (*c*) levels with respect to the global ROM predynamic (Pre-D) and postdynamic (Post-D) loadings

^{121006-4 /} Vol. 138, DECEMBER 2016



Fig. 3 Mean (\pm 95% Cl) rotation angle (flexion positive) at Occ/C1 (*a*), C1/C2 (*b*), and C2/C3 (*c*) levels during dynamic flexion, and dynamic extension (*d*)–(*f*) applied at 100 deg/s to the C3 level from 20 to 81 ms

actuation stopped; the reverse was observed in dynamic extension loading. Some specimens were excluded from the statistical comparisons, since no C2/C3 rotations during quasi-static loading were within 0.1 deg of the C2/C3 rotations of dynamic loading (Table 3). The maximum posterior–anterior sliding was -1.07 ± 0.42 mm in dynamic flexion and 1.13 ± 0.43 mm in dynamic extension. The separation–compression was less

consistent, but remained within 0.53 mm of separation and 0.61 mm of compression relative to the neutral position for all the specimens in all the tests. The maximum linear strain across the facet was $3.92 \pm 1.33\%$ in dynamic flexion and $2.58 \pm 0.98\%$ in dynamic extension. Dynamic loading produced significantly more posterior–anterior sliding than quasi-static loading ($p \le 0.043$) (Table 3). The maximum linear strain across the facet joint was



Fig. 4 Mean (\pm 95% Cl) peak MPS ((*a*) and (*b*)) and peak SS ((*c*) and (*d*)) of the C2/C3 facet capsule during dynamic actuation of the C3 applied at 100 deg/s from 20 to 8 ms in flexion ((*a*) and (*c*)) and extension ((*b*) and (*d*))

Journal of Biomechanical Engineering



Fig. 5 Mean (\pm 95% CI) peak MPS (*a*) and mean MPS (*b*) on the C2/C3 facet capsule during flexion and extension at 100 deg/s compared to quasi-static loading. An asterisk (*) denotes a significant difference (p<0.05).

lower than the peak MPS from the full-field capsular analysis, and no differences were found between the linear facet strain in either dynamic flexion or dynamic extension compared to quasi-static loading ($p \ge 0.345$) (Table 3).

4 Discussion

This study compares the cervical spine kinematics, joint kinematics and strains, and full-field strain of the C2/C3 facet capsule during dynamic flexion–extension with corresponding physiological ROMs. The dynamic loading profiles simulate those sustained by the neck of a seated occupant during rear-end and frontal collisions [49]. Although the vertebral rotations during dynamic loading did not exceed physiological ROMs (Table 1) [50,51], both the MPS and SS in the facet capsule were increased during dynamic flexion relative to those sustained during corresponding quasi-static rotations (Fig. 5; Table 2). The 100 deg/s rotation rate used in the present study was selected to replicate the rotation of the C3 during a low-speed collision [49,52], from which neck injury can commonly result. Further, the acceleration of loading (1.8–2.8 g) was in line with the previously reported data [25,49].

Muscle activity was not simulated during this study, although this is likely to have minimally affected the results. Muscle activity has been shown to be limited to an activation response time of approximately 51 ms and the time to maximum force production

for neck muscles is approximately 114 ms [53], the time periods over which the C3 vertebra was actuated (61 ms) and over which data were analyzed (130 ms) would not incorporate muscle contributions. The intervertebral rotations during dynamic flexion were comparable to those previously reported (Table 4). The relatively large rotations, particularly in extension reported by Grauer et al. [54], may be the result of the flexion moment due to the mass of the head being was counteracted by a pneumatic suspension system. The study of Ivancic et al. [39] used muscle force simulation to stabilize the head, which resulted in similar peak flexion and extension rotations to the present study. While a stabilizing preload could have been applied to simulate passive muscle activity, it was not used in the present study. Although this is a limitation as the stability that muscle forces would provide in vivo was absent in these experimental conditions, the results of this study could be taken as a worst-case scenario of the neck response to front- and rear-end collisions. Further research into the effect of muscle forces on the cervical spine and facet strain responses during dynamic loading exposures would help to assess the likelihood of injury under different loading conditions.

The facet kinematics in this study exhibit similar behavior to those previously reported, with the actuation of the C3 vertebra causing posterior sliding of the C2 level relative to the actuated level [25,49,55]. However, the greatest sliding during rear-end collision simulation in our study was anterior sliding (Table 3),

Table 1 Mean \pm standard deviation (SD) maximum and minimum rotations (flexion positive) at each cervical spinal level during dynamic flexion and extension, and quasi-static rotation

Level	Flexion		Extension		Quasi-static	
	Max	Min	Max	Min	Max	Min
Occ/C1	5.31 ± 2.76	-0.39 ± 0.37	0.80 ± 1.04	-7.73 ± 3.30	5.20 ± 2.92	-13.11 ± 4.78
C1/C2	7.88 ± 3.53	-0.02 ± 0.03	0.96 ± 1.20	-1.81 ± 0.54	14.60 ± 5.26	-3.27 ± 1.52
C2/C3	1.74 ± 1.14	-1.54 ± 0.98	2.04 ± 2.08	-2.17 ± 0.65	3.71 ± 4.40	-4.06 ± 3.06

Table 2 Mean \pm SD SS in the C2/C3 facet capsule during dynamic flexion and extension exposures and equivalent C2/C3 rotation during quasi-static loading

Test	Parameter	п	100 deg/s	Quasi-static	Significance
Flexion	Maximum SS (%)	5	5.34 ± 1.60	1.73 ± 2.20	0.028 ^a
	Minimum SS (%)	6	-4.30 ± 2.73	-4.69 ± 5.55	0.917
Extension	Maximum SS (%) Minimum SS (%)	5 6	2.73 ± 1.83 -4.38 ± 1.50	2.43 ± 2.10 -3.37 ± 5.49	0.028 ^a 0.463

^aA significant difference between dynamic and quasi-static tests (p < 0.05).

121006-6 / Vol. 138, DECEMBER 2016

Table 3 Mean ± SD maximum facet joint sliding, separation, compression, and linear strain

Test	Parameter	n	100 deg/s	Quasi-static	Significance
Flexion	Posterior-anterior sliding (mm)	5	-1.01 ± 0.46	-0.35 ± 0.31	0.043 ^a
	Separation (positive) (mm)	6	0.20 ± 0.18	0.18 ± 0.19	0.600
	Compression (negative) (mm)	6	-0.22 ± 0.21	-0.08 ± 0.17	0.027 ^a
	Maximum strain across the facet (%)	5	3.92 ± 1.48	3.38 ± 2.56	0.345
Extension	Posterior-anterior sliding (mm)	5	1.19 ± 0.46	0.44 ± 0.56	0.042 ^a
	Separation (positive) (mm)	5	0.22 ± 0.09	0.16 ± 0.19	0.416
	Compression (negative) (mm)	5	-0.24 ± 0.11	-0.18 ± 0.18	0.343
	Maximum strain across the facet (%)	4	2.78 ± 1.18	2.56 ± 2.30	1.000

^aA significant difference between dynamic and quasi-static tests (p < 0.05).

which occurred after the actuation of C3 had ceased. Although most prior work generating quantitative data has focused on the initial exposure period and the return of the spine to the neutral posture [25], it is possible that the kinematics immediately following a dynamic exposure may actually be injurious if the magnitude of deceleration is sufficient. Further, the facet compression and separation in our studies were within 0.61 mm of the neutral position in all the tests. The peak facet compression during rearend collision simulation of -0.22 ± 0.21 mm was lower than the -0.9 to 2.8 mm previously reported for 3.5–4.4 g impacts [25,28,38]. However, despite being small, the compressions in the current study were significantly greater than during the quasistatic tests at an equivalent intervertebral rotation, which has not been previously reported. This suggests that while the present testing may underestimate the absolute compression, an increased risk of an injury may exist.

Although there was little variation across the age of the specimens (mean of 66 years), it is possible that younger specimens would exhibit different mechanical and physiological responses under the loading conditions of the present study. However, the specimens tested represent a realistic population that may be subjected to such loading and are of a similar age to those tested in the majority of in vitro cadaveric studies.

The dynamic loading exposures did not alter the overall kinematics during quasi-static loading (Fig. 2). Rear-end impact simulations using both incremental accelerations of 2, 3.5, 5, 6.5, and 8 g and a single 8 g exposure increase both the ROM and the neutral zone, suggesting that soft tissue injury may be induced [56].



Fig. 6 Representative full-field strain for specimen 1 ((*a*) and (*b*)) and specimen 4 ((*c*) and (*d*)). The peak MPS during dynamic flexion ((*a*) and (*c*)) and the MPS at equivalent C2/C3 rotation during quasi-static testing ((*b*) and (*d*)) are shown. The variable MPS across the facet capsule was determined; the arrows show the MPS direction within each element.

Journal of Biomechanical Engineering

Table 4 Mean \pm standard deviation (SD) maximum and minimum rotations (flexion positive) at each cervical spinal level during dynamic flexion for the current study in comparison to published cadaveric simulations of rear-end collisions

Level	Present study (2.8 g)		Ivancic et al. (3.5 g) [39]		Grauer et al. (2.5 g) [54]	
	Max	Min	Max	Min	Max	Min
Occ/C1 C1/C2 C2/C3	5.31 ± 2.76 7.88 ± 3.53 1.74 ± 1.14	-0.39 ± 0.37 -0.02 ± 0.03 -1.54 ± 0.98	4.9 ± 3.7 2.0 ± 1.6 1.5 ± 1.1	-0.8 ± 1.1 -1.8 ± 2.1 -4.9 ± 3.9	9.2 ± 5.8 16.6 ± 8.7 1.7 ± 1.1	-13.4 ± 7.5 -13.8 ± 3.6 -7.0 ± 2.6

However, while the current study only assessed the rotation at each vertebral level with respect to the global ROM in quasi-static tests that were performed before and after dynamic loading, the fact that there were no significant differences in the vertebral rotations indicates that soft tissue damage is not likely to have occurred in this study. Moreover, while it is possible that such an approach would not identify soft tissue damage as accurately as would be evident by a change in neutral zone, it is also possible that the relatively low accelerations and decelerations of 1.8–2.8 g used in dynamic exposures of the current study would not be sufficient to elicit such a change.

The inertia of the specimen when the actuation ceased led to a brief period when the specimens adopted an S-shape (Fig. 7). This inertial effect in dynamic flexion corresponded approximately to the point of the peak MPS, maximum SS, and anterior facet sliding (Fig. 4), all of which were significantly higher compared to quasi-static loading (Fig. 5; Tables 2 and 3). This inertial effect on the upper cervical spine indicates that the deceleration following an impact is critical in terms of facet mechanics relating to the peak strain, and that minimizing the deceleration following a rearend collision would limit the combination of C2/C3 rotation and anterior sliding that may lead to injury and pain. Previous studies have reported that the induction of an S-shape to the cervical spine and altered vertebral centers of rotation and facet kinematics are likely to be critical in the development of injury [25,28,36]. The present results are consistent with this hypothesis, which demonstrate that the peak facet sliding corresponds with the peak MPS and SS. The altered kinematics combined with increased facet capsule strain observed in the dynamic flexion tests compared to quasi-static loading suggest that loading to C2/C3 is different in those two scenarios. The present study compares peak dynamic strains with quasi-static strains at equivalent C2/C3 rotations; findings suggest that the significant differences observed here may be related to altered loading which causes increased facet sliding and increases the likelihood of facet capsular ligament injury. However, other loading scenarios and their effects on the spinal

kinematics and full-field facet capsule strains should be investigated especially compared to physiological loading in order to fully understand how injuries may occur in difference collision scenarios.

Although showing similar kinematic patterns relating to the inertial effect of the surrogate head as in dynamic flexion, dynamic extension simulating a frontal collision did not produce increased MPS or SS following actuation (Fig. 4; Table 2). In contrast, the maximum shear strain, which occurred during actuation (20–81 ms), was significantly higher than during quasi-static loading (Table 2). This may be due to the center of mass of the head naturally falling into flexion, and thus limiting the extent to which the dynamic loading increased the MPS and SS. It is possible that by implementing a follower-load to simulate passive muscle forces and to maintain the neutral position of the head, the MPS and SS would increase once the actuation stopped, similar to what was observed in the rear-end collision simulations of the present study (Fig. 4).

The full-field strains showed that both MPS and SS were inhomogeneous (Fig. 6), and that the full-field strain measurement of the capsule more accurately identifies local responses within the facet capsule compared to measuring the linear strain across the entire facet joint. The adoption of the previously published method to assess overall facet behavior [25] did not reveal the significance in terms of increased strain during dynamic flexion (Table 3) that was identified using the full-field strain method (Fig. 5). These results are consistent with the previous study [25] but emphasize that measuring the linear strain across the entire facet may underestimate the strain magnitude. The peak MPS during dynamic flexion $(12.52 \pm 4.59\%)$ approached the peak MPS that has been reported in in vivo studies of facet capsule stretch in a rat model of facet-mediated painful mechanical injury [35]. Interestingly, the linear strains measured across the entire facet joint in the current study during dynamic flexion (Table 3) are similar to the peak MPS induced in a nonpainful in vivo group of Dong et al. (Table 5). The peak MPS measured during dynamic



Fig. 7 Mean rotation angle (flexion positive) during dynamic actuation applied at 100 deg/s to the C3 level from 20 to 81 ms in either flexion (a) or extension (b). The visual representations above the plots show the shape of the spine at 0, 40, 81, 120, and 150 ms.

121006-8 / Vol. 138, DECEMBER 2016

Table 5 Summary of mean \pm SD facet strain data from painful and nonpainful in vivo studies and during quasi-static (QS) and dynamic In vitro loading

Study	Measurement	Test method	Level	Variable	Strain (%)	
Dong et al. [35]	2D full-field strain (MPS)	In vivo rat facet retraction	C6/C7	Painful	19.43 ± 11.43 6 29 + 2 64	
Present study	3D full-field strain (MPS)	In vitro rear-end collision	C2/C3	2.8 g	12.52 ± 4.59 7.63 ± 4.35	
	Linear strain of facet	In vitro rear-end collision	C2/C3	2.8 g	3.92 ± 1.33 3.92 ± 2.56	
Pearson et al. [25]	Linear strain of facet	In vitro whiplash	C2/C3	3.5 g	3.38 ± 2.30 13.4 ± 9.3 0.4 ± 7.1	
Panjabi et al. [31]	Linear strain of facet	In vitro whiplash	C2/C3	QS 2.5 g	9.4 ± 7.1 15.9 ± 20.0	

flexion in the present study is similar to the strain in the C2/C3 facet capsule during whiplash exposures to human cadaveric cervical spines [25,31] (Table 5). However, since the peak MPS measured from full-field strain analysis was much higher than the linear strain across the facet during both dynamic flexion and extension in the present study (Table 3), it is likely that the latter method will fail to identify the peak strain, and may, therefore, underestimate the likelihood of injury, pain, and/or other pathophysiological dysfunctional responses due to a traumatic exposure.

Pain can be induced at facet capsule strains much lower than the subcatastrophic failure strains that have been reported during whiplash at approximately 35-67% [32,40,41]. The notion that a threshold for pain production is lower than failure may be due to microscopic injury of the collagen fiber matrix [42,43,57,58]. The onset of injury at the microstructural level has been identified using quantitative polarized light imaging [42,43,57] and using proxies such as afferents in the ligament [59-61]. The full-field strain analysis of the present study does show that the peak MPS from dynamic flexion generally occurs in the anterior aspect of the facet capsule and is aligned along the facet joint (Fig. 6), which is consistent with increased anterior sliding at the time of peak MPS. Little correlation was found between fiber realignment and the maximum MPS or SS in isolated capsules loaded in distraction [42,57], or posterior retraction [43]. It is possible that higher resolution full-field strain analysis would better predict the magnitude and location of capsule injury, but this method would not account for subsurface injury of the capsule that identified capsule damage through collagen fiber realignment using QPLI.

The full-field shear strain in the cervical facet capsules (Fig. 4) has not previously been investigated in dynamic loading exposures representative of either frontal or rear-end collisions in multilevel cervical specimens; yet, it has been measured during the retraction [43,62] and distraction [42] of isolated cervical facet joints, and in functional spinal units undergoing pure moment testing [32]. There is limited shear strain data for the facet capsule at subcatastrophic or catastrophic failure, making it difficult to assess the maximum and minimum shear strain of the present study (Table 2) in relation to either injury or pain. The maximum and minimum SS in the present study were lower than measured during quasi-static loading [42,43]; but this may be due to both the direction and rate of loading, which affect the location, magnitude, and orientation of both MPS and SS. The adoption of methods to measure local ligament kinematics and kinetics along with the facet joint kinematics measured in the present study would provide a greater understanding of how collagen fiber responses and strains relate to capsule injury under loading conditions representative of traffic collisions and would assist in understanding which conditions increase the likelihood of injury.

Full-field strain measurements provide a quantitative measure of the inhomogeneous strain behavior of the facet capsule, which allows a more accurate understanding of the risk of injury during dynamic loading compared to linear strain analysis across the entire facet. The present study demonstrates that loading

Journal of Biomechanical Engineering

representative of low-speed rear-end collisions produces injurious MPS in the C2/C3 facet capsule, and that such exposures alter the anterior–posterior sliding motions of the facet joint and increase the facet capsular MPS and SS relative to those during physiologic flexion–extension. Although upper cervical spine loading representative of low-speed frontal collisions leads to similar changes in anterior–posterior joint sliding and SS in the facet capsule compared to quasi-static flexion–extension, those loading conditions did not lead to increased MPS in the C2/C3 facet capsule.

Currently, there are no published reports on whether there are level-by-level differences in the anatomy and neurophysiology of the cervical spine that would lead to spinal level-specific facet capsule injury thresholds under dynamic loading conditions. Additional in vivo studies could provide deeper understanding of whether there are potential differences in strain and/or pain thresholds for the facet capsule or different levels of the cervical spine. Furthermore, it remains difficult to identify the microstructural injury mechanisms within the facet during dynamic loading conditions representative of in vivo scenarios, such as traffic collisions. But the significantly elevated strains under dynamic exposures compared to quasi-static loading in the present study suggest that strains in the C2/C3 facet capsule may reach levels sufficient to induce pain and/or other local injury. Future work combining these data with microstructural analyses of isolated facet capsules and in vivo studies could help fully define how such loading exposures occur, and what loading conditions put the cervical spine at the greatest risk of injury. The results also demonstrate that the deceleration following a rear-end collision is just as critical in avoiding injury as the collision itself, and the ability to minimize such decelerations through improved safety mechanisms may reduce the likelihood of neck pain following such exposures.

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