INVESTIGATION OF FACET JOINT RESPONSE UNDER REAR IMPACT CONDITIONS USING FE MODEL OF THE CERVICAL SPINE

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ABSTRACT

Whiplash injury resulting from rear impact is a significant issue in terms of societal cost, and the resulting pain and reduction in quality of life. The facet joints in the cervical spine have been identified as a source of pain in whiplash injuries; however, the responses of these joints are difficult to measure in vivo or in vitro. In this study, a detailed explicit FE model of the cervical spine was used to investigate facet joint response under rear impact loading conditions. The model represents a mid-size male with detailed vertebrae, discs, ligaments and Hill-type active muscles. This model was previously validated extensively at the segment level and validated for frontal impact scenarios. In this study, the cervical spine model was validated against rear impact volunteer and cadaver tests (13 volunteers exposed to 28 rear impacts at speeds of 5 to 7kph; 26 cadavers exposed to rear impacts at speeds of 5 to 15.5kph) using experimental acceleration, displacement and rotation traces of the T1. Capsular ligament (CL) strains were measured in the model and compared to values presented in the literature to identify pain or sub-catastrophic failure. Simulation of 4, 7, and 10g rear impacts showed good agreement with the experimental data. The predicted CL strains were below or near the approximate threshold for pain and sub-catastrophic damage (35% strain), and exceeded this value for a 12g rear impact case. This study included muscle activation, and provides a link between published strain limits for facet joint capsules evaluated in controlled lab conditions and strains predicted under rear impact loadings.

INTRODUCTION

Whiplash, or cervical spine injury resulting from low speed rear impact, is a significant issue with the annual societal costs in the United States estimated to be between 4.5 and 29 billion dollars (Keinberger 2000, Freeman 1997). Whiplash can reduce the victim's quality of life for a significant amount of time, as up to 33% of patients continued to seek treatment for whiplash 33 months after sustaining the injury (Freeman 1997). With respect to sources of pain, one focus of whiplash research has been on capsular ligament (CL) strain with several clinical and biomechanical studies implicating this anatomical site as a likely source of injury (Lord, et al. 1996, Barnsley, et al. 1995, Lu, et al. 2005, Lee, Davis, et al. 2004, Lee, Franklin, et al. 2006, Ivancic, et al. 2008, Quinn, et al. 2007, Panjabi, et al. 1998, Pearson, et al. 2004, Deng 1999a). Clinical studies using double-blind anesthetic blocks have shown that 54% to 60% of whiplash patients have CL pain (Barnsley, et al. 1995, Lord, et al. 1996). By measuring behavioral sensitivities or nerve discharge, in-vivo animal models of the goat and rat have shown that tensile force applied across the facet joint led to pain (Lu, et al. 2005, Lee, Davis, et al. 2004). In the rat model, it was shown that in-vitro sub-catastrophic failure of the CL occurred at a distraction magnitude of 0.57mm that led to pain for up to 14 days in-vivo (Lee, Davis, et al. 2004, Lee, Franklin, et al. 2006, Quinn, et al. 2007). Authors have shown that CL strain in cadavers and cadaveric cervical spine sections peak values range from 28.5% to 39.9% during low speed rear impact, which exceeds physiologic strain of this structure (Panjabi, et al. 1998, Pearson, et al. 2004, Deng 1999a). The stiffness of CLs exposed to rear impact was less than the control CLs, which showed that some damage had occurred in the ligaments despite a lack of visual indication (Ivancic, et al. 2008).

Four types of studies have been undertaken to measure or predict the level of strain in the CL during rear impact scenarios: full-body cadaver sled tests, full cervical spine bench-top sled tests, quasi-static cervical spine motion segment tests, and finally, computational models. These different approaches have provided important information and understanding, but with some limitations. Deng et al. (1999a) performed a series of 26 rear-impact sled tests on six post-mortem human subjects and

measured capsular strain using a pair of lead spheres in each vertebra, which were tracked by high-speed x-ray. However, one of the challenges with cadaver studies is the lack of active musculature, which affects the kinematics of the neck (Thunnissen, et al. 1995, van der Horst, et al. 1997). Studies that use a bench-top sled to impose rear-impact loads on cervical spine sections (Panjabi, et al. 1998, Pearson, et al. 2004), typically do not include the upward motion and extension of the T1 caused by straightening of the spine and the torso ramping up the seat (Deng 1999a). Winkelstein et al. (1999) and Seigmund et al. (2000) applied bending and shear loads to isolated motion segments, and measured the capsular strains. However, this type of study did not include dynamic effects, such as changes to the axis of rotation of each vertebra during rear-impact (Ono, et al. 1997). Current computational models that have been used to calculate CL strain under rear impact include the TNO model (Stemper, Yoganandan and Pintar 2005) and THUMS model (Kitagawa, Yasuki and Hasegawa 2008), but these studies did not incorporate active musculature or detailed facet joints.

This current study is based on a detailed validated finite element model of the cervical spine (Panzer 2006, Panzer and Cronin 2009) and the prediction of capsular ligament strains during rear impact. These strains were compared to published limits for sub-catastrophic failure and pain. The CL strains were measured at every cervical level at the anterior and posterior position of each facet joint. The results of this study are unique in that active musculature and realistic loadings were included.

METHODS

The finite element model used in this study represents a mid-size male and was developed at the University of Waterloo (UW); a full description is available in Panzer (2006) (Figure 1). The model was developed with the focus on accurate geometric and material representation at the local tissue level. The vertebrae geometry was based on the model developed by Y. C. Deng et al. (1999b) and the vertebrae were modeled as rigid for computational efficiency. The intervertebral discs were constructed with solid elements for the annulus fibrosus ground substance and layers of shell elements representing the fibre lamina, and solid elements to model the nucleus pulposus. The facet joints were modeled with a superior and inferior layer of solid elements for the articular cartilage with a squeeze-film model to simulate the synovial fluid (Figure 2). Ligaments were represented using multiple 1D non-linear spring elements. In total, 90 pairs of active Hill-type 1D elements were used to model 27 muscle pairs in the cervical spine. Both the flexors and extensors were activated 74ms after impact (Siegmund, Sanderson, et al. 2003). Studies have found that flexor and extensor muscles activate at the same time and that EMG muscle signals start at 60 to 79ms after impact (Siegmund, Sanderson, et al. 2003, Ono, et al. 1997, Roberts, et al. 2002, Szabo and Welcher 1996). It is possible that the actual muscle activation scheme for rear impact is more complex, but no conclusive information is available at this time. The material models for all the components were based on studies in the literature. Viscoelasticity and anisotropy were incorporated where applicable.

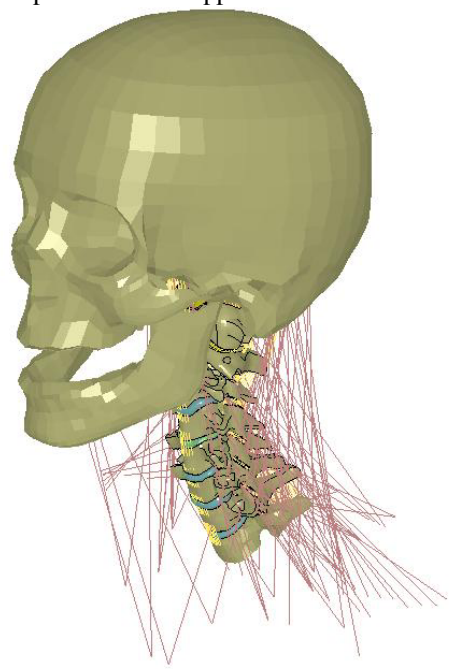


Figure 1. FE model of the cervical spine. Pink lines represent muscle elements, and yellow lines represent ligaments.

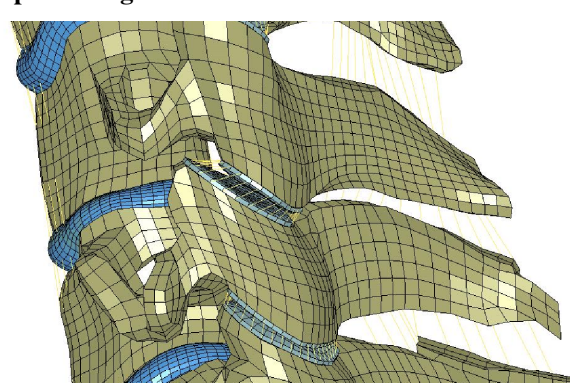


Figure 2. Facet joint detail, showing the CL and articular cartilage.

The cervical spine model used in this study was previously validated at the segment level in flexion, extension, lateral bending, axial rotation, tension, compression, and anterior, posterior, and lateral shear (Panzer 2006). Panzer found generally excellent correlation (within one standard deviation) with the quasi-static loadings. This model was validated against volunteer frontal impacts up to 15g (Panzer 2006).

For the model to be used to predict CL strains during rear-impact, the model must first be validated for this type of loading. Davidsson et al. (1998) and Deng (1999a) were found the most suitable based on the severity of impact, number of test samples, and full data of the T1 motion in the sagittal plane. The head kinematic response corridors and T1 inputs for Davidsson's experiments came from Hynd et al. (2007).

Davidsson et al. (1998) performed 28 rear impacts on thirteen human volunteers at speeds between 5 and 7kph with an average peak acceleration of 3.6g. The test involved the collision of a bullet sled with a stationary target sled, which seated a volunteer on a laboratory seat with a headrest. To model these rear impacts, the average T1 X-acceleration (fore-aft), Z-displacement (superior-inferior), and Y-rotation (flexion-extension) were input into the cervical spine model T1 as prescribed motion constraints (Figure 3). The model's T1 was constrained in all other directions, and the head was not constrained. The headrest for the test was described as a stiff backing plate covered with 4cm of foam, attached to a rigid frame with four coil springs of a specified preload and stiffness. The mass of the headrest and dimensions were also specified. To model the headrest, the average sled x-acceleration was input as prescribed motion to the frame attachment points of the springs (Figure 3). The headrest material properties were non-linear viscoelastic based on automotive seat cushion material tested at UW and the stiff backing was assumed to be pine with orthotropic elastic material properties (Green, Winandy and Kretschmann 1999, Cambell and Cronin 2007). Muscle activation was included for validation against Davidsson et al. (1998) to mimic the behavior of volunteers.

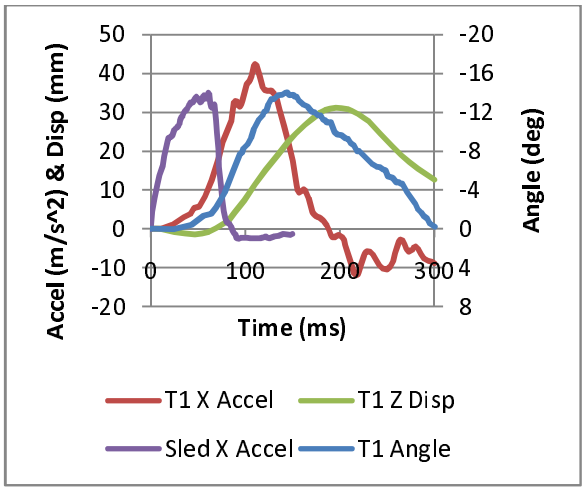


Figure 3. Davidsson et al. (1998) T1 Inputs for a 4g loading case, X direction is positive forward, Z direction is positive upwards, and T1 rotation is positive in extension. Inputs are in a fixed global coordinate system.

Deng (1999a) performed a series of 26 rear impacts on 6 whole body cadavers at delta velocities ranging from 5 to 15.5kph, and accelerations from 5 to 9.9g. These experiments involved a cadaver seated in a custom seat, with or without a headrest, and being accelerated from rest using a pneumatic cylinder. When modeling these tests the headrest was not included, because in the experiments the headrest was initially positioned at least 100mm away from the cadavers head and did not influence kinematics until late in the simulation, and the author found all peak CL strains occurred before headrest contact (Deng 1999a). Two specific runs were chosen to simulate based on the impact severity and available data. The experimental T1 X-acceleration, Z-acceleration, and Y-rotation were input into the model as prescribed motion (Figure 4). The T1 was constrained from moving in any other direction and everything else was free in all directions. Muscle activation was not included in the validation against Deng (1999a) to mimic the behavior of cadavers. More aggressive rear impacts were undertaken by scaling the 10g simulation to 11 to 20g to identify the threshold for CL strain injury in the model.

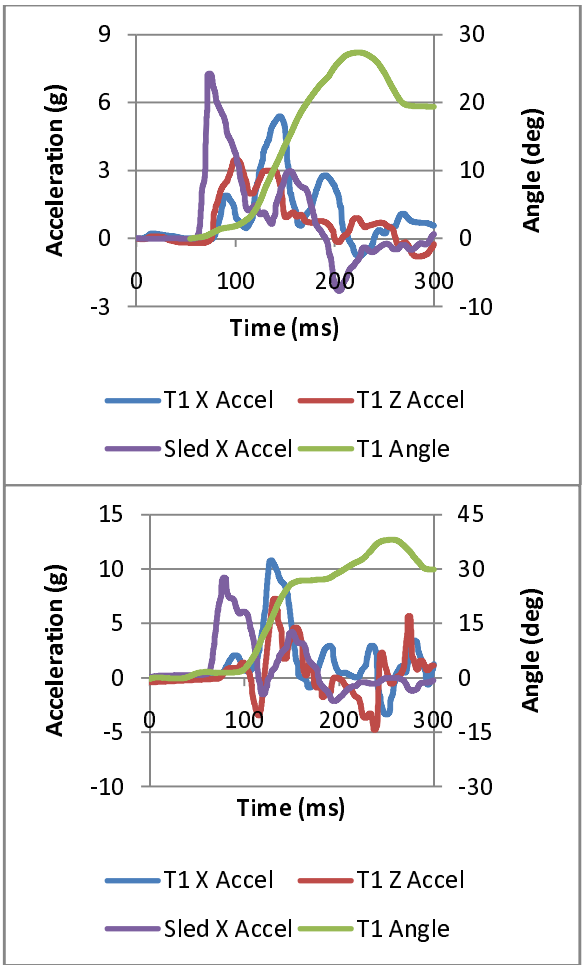


Figure 4. Deng (1999a) T1 Inputs for 7g (above) and 10g (below) loading cases, X direction is positive forward, Z direction is positive upwards, and T1 rotation is positive in extension. The sled acceleration was not used in the simulation. Inputs given in a rotating anatomical coordinate system.

The CL strains in the model were calculated by measuring the change in length of 1D discrete elements representing the CL and dividing by CL ligament lengths measured on cadavers (Panjabi, et al. 1998). The strains were calculated at the anterior most position and posterior most position of the facet joint for each cervical level. These positions were chosen, because the CL strain should be the most extreme at these locations.

RESULTS

The model was able to reproduce the kinematics of volunteers in Davidsson's (1998) experiment (Figures 5 to 9). The head centre of gravity (CG) motion in the fore-aft (X) direction was in excellent agreement with the data for both with respect to the first thoracic vertebrae (T1) and with respect to the sled (Figures 5 & 6). Head rotation in the sagittal plane with respect to the T1 was in good agreement with the volunteer data (Figure 7). The occipital condoyle (OC) superior-inferior (Z) direction movement was also in good agreement with the data, when measured with respect to the sled and with respect to the T1 (Figures 8 & 9).

The model's response to a 7g rear impact was in reasonable agreement to the cadaver tests performed by Deng (1999a) (Figures 10 to 13). The rotations of the model's upper cervical spine joints (C1-C2 and C2-C3) did not exhibit enough relative flexion, and the middle cervical spine joint (C4-C5) did not exhibit enough relative extension (Figure 10). The combination of relative vertebral rotations lead a head rotation that was a good fit to Deng's data (Figure 11). The model's head CG X-acceleration has a similar double peak shape to the cadaver results, but the peaks were on the lower end of the measured data (Figure 12). In the Z-direction the model's acceleration did have similar peaks or shape compared to the cadaver data (Figure 13).

At 10g, the model's responses were a reasonable fit to a limited data set from Deng (1999a) (Figures 14, 15, & 16). The angle of the model's head was a good fit to cadaver head angles at similar impact accelerations (Figure 14). The X-acceleration of the head CG was of similar shape, and the peak accelerations lie mostly in the data spread of cadaver results (Figure 15). As was the case for the 7g impact, the head's CG Z-acceleration was neither a similar shape nor did it have similar peak values when compared to the cadaver data (Figure 16).

When considering the CL strains in the different load cases, the CL strain on the anterior portion of the C5-C6 facet joint was always greater than the posterior portion, because the motion segments were loaded in extension and posterior shear (Figures 17 to 20). The model predicted a peak CL strain of 22.6% at the C4-C5 level during the 4g rear impact (Table 1). The next highest strains were 28.6% and 32.4% at the C5-C6 and C2-C3 levels respectively during a 10g impact with passive muscles (Table 1). When including active muscles, the CL strain reduced at every level except for C4-C5. A CL strain of 35.4% was measured in the C2-C3 during a 12g rear impact, which exceeded the sub-catastrophic strain of the CL ligament reported by Seigmund et al. (2001) (Table 1).

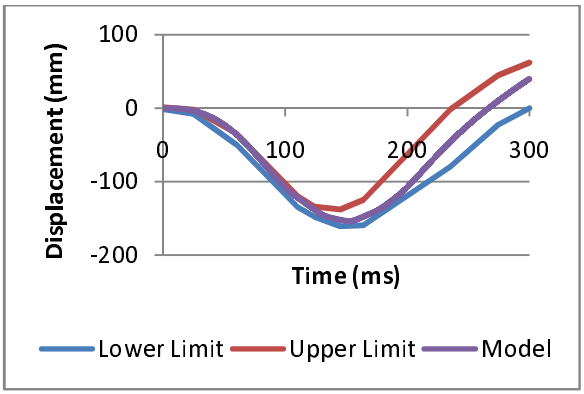


Figure 5. 4g - Head CG X-Displacement w.r.t. the Sled (Davidsson, et al. 1998).

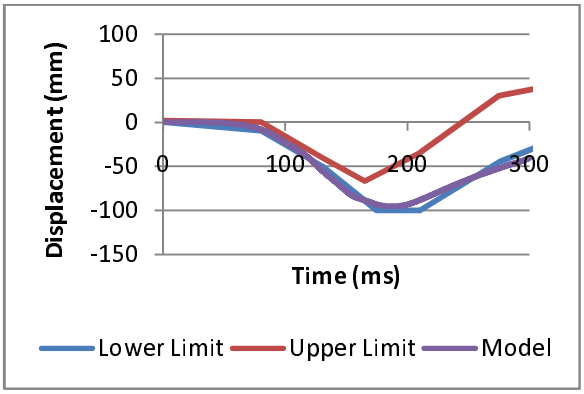


Figure 6. 4g - Head CG X-Displacement w.r.t. the T1 (Davidsson, et al. 1998).

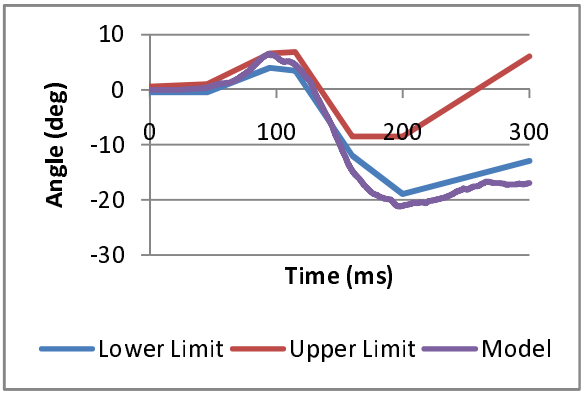


Figure 7. 4g - Head rotation w.r.t. the T1 (Davidsson, et al. 1998).

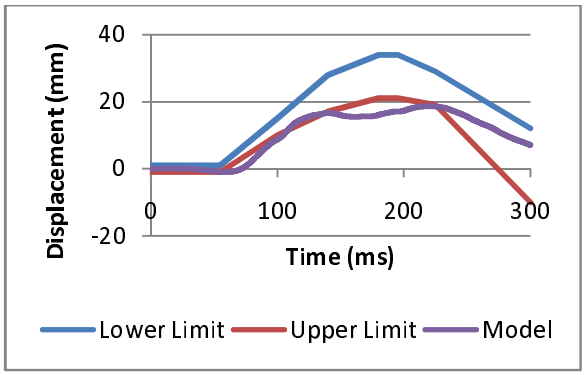


Figure 8. 4g - O.C. Z-Displacement w.r.t. the Sled (Davidsson, et al. 1998).

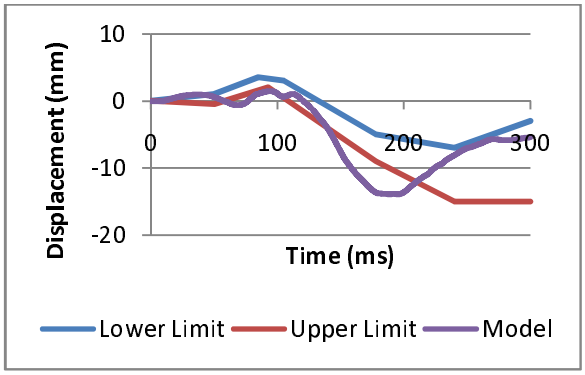
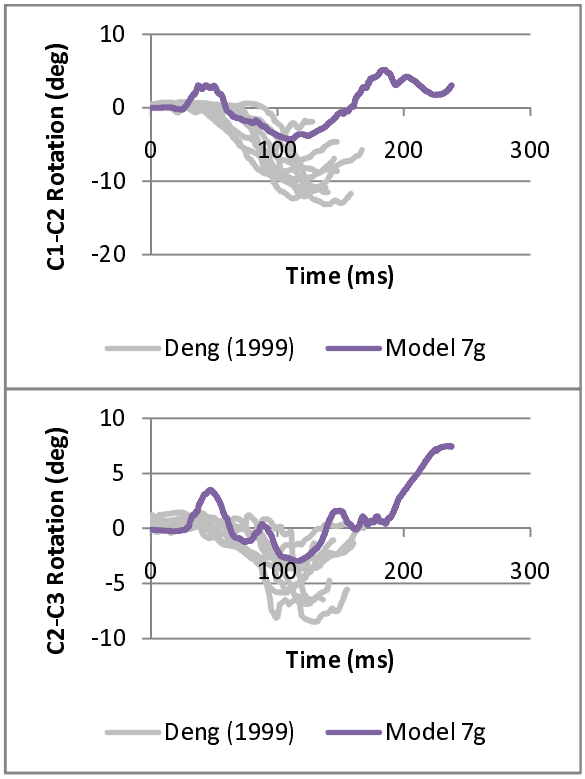


Figure 9. 4g - O.C. Z-Displacement w.r.t. the T1 (Davidsson, et al. 1998).



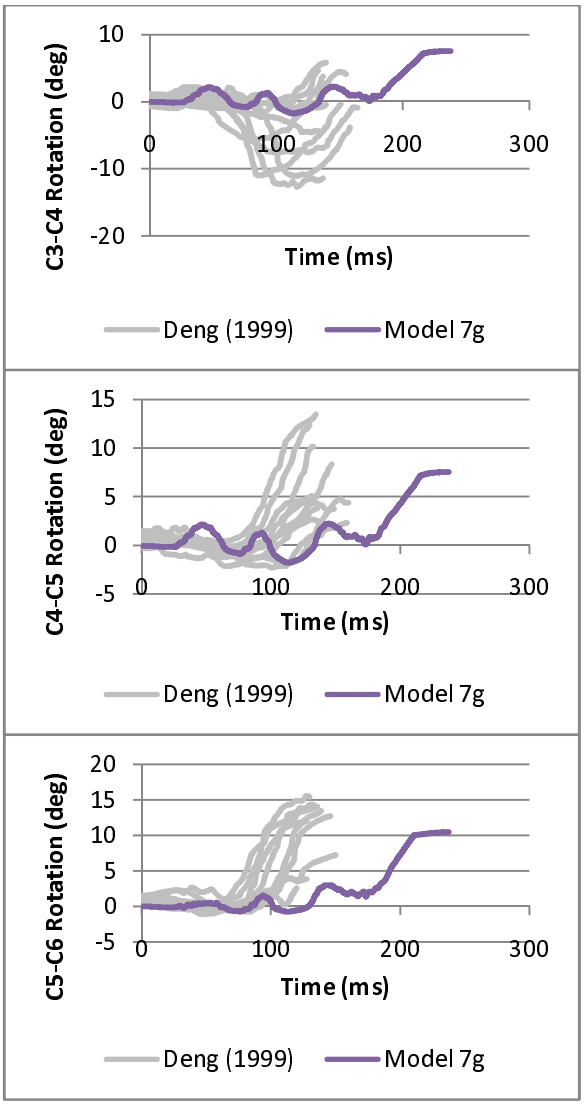


Figure 10. Spine segment relative rotations in the sagittal plane. Positive angles for extension.

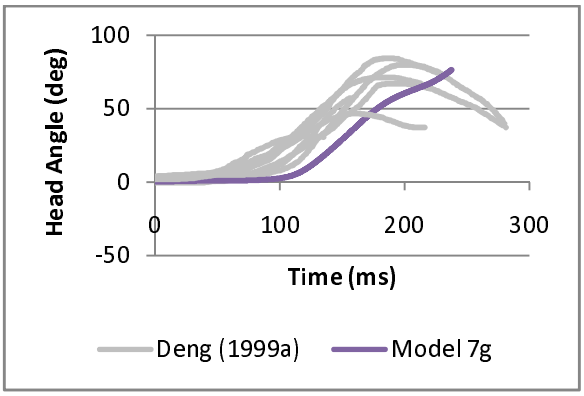


Figure 11. Cadaver head rotations for loadings between 6 to 8g compared to the model's response at 7g. Positive angles for extension.

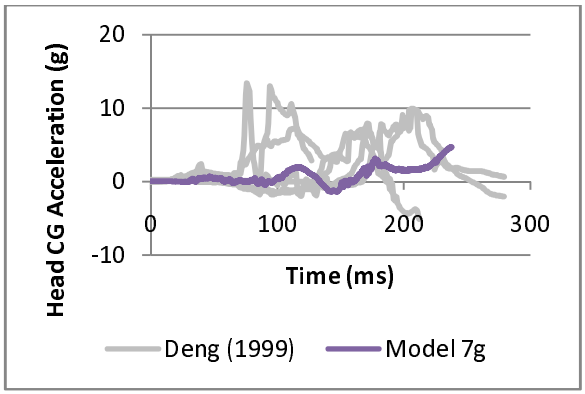


Figure 12. Cadaver head CG X-acceleration for loadings between 6 to 8g compared to the model's response at 7g. Positive g's in the forward direction.

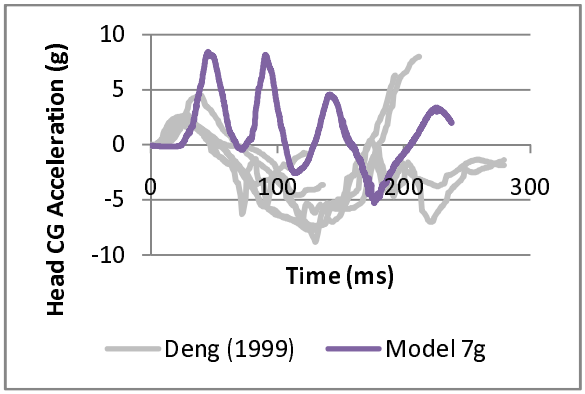


Figure 13. Cadaver head CG Z-acceleration for loadings between 6 to 8g compared to the model's response at 7g. Positive g's in the upward direction.

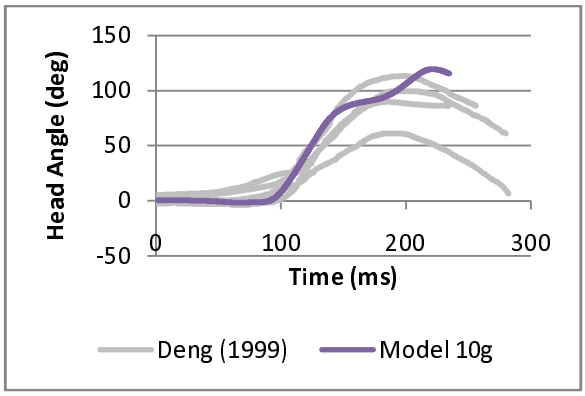


Figure 14. Cadaver head rotations for loadings between 8 to 10g compared to the model's response at 10g. Positive angles for extension.

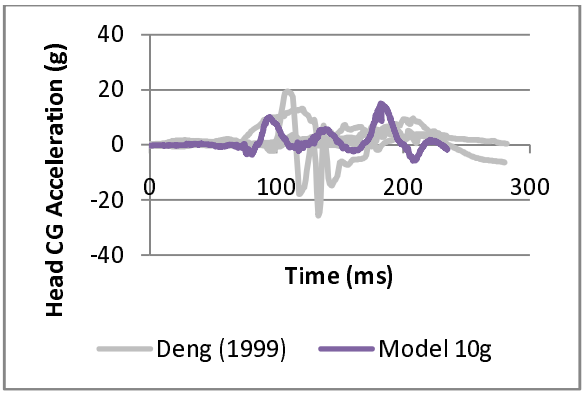


Figure 15. Cadaver head CG X-acceleration for loadings between 8 to 10g compared to the model's response at 10g. Positive g's in the forward direction.

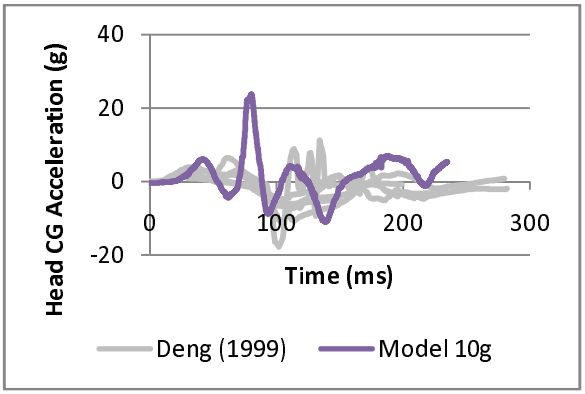


Figure 16. Cadaver head CG z-acceleration for loadings between 8 to 10g compared to the model's response at 10g. Positive g's in the upward direction.

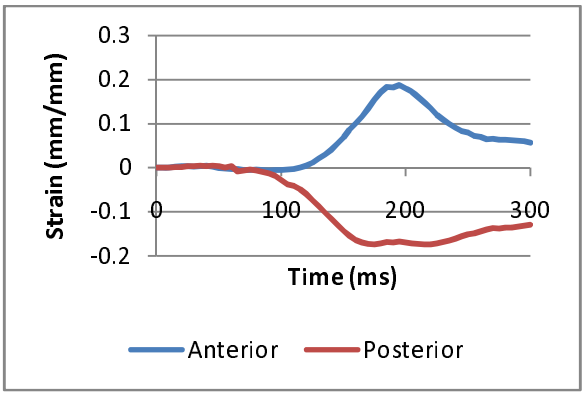


Figure 17. Predicted C5-C6 CL strain for a 4g impact.

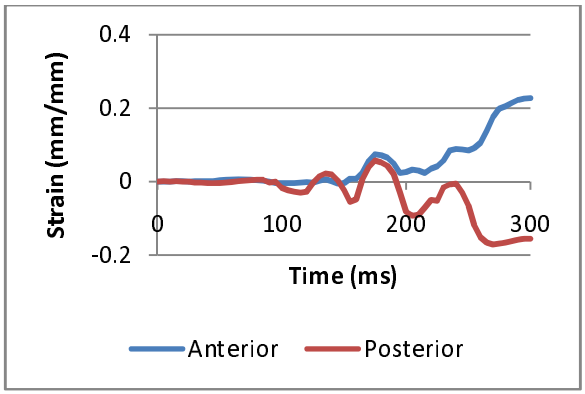


Figure 18. Predicted C5-C6 CL strain for a 7g impact with passive muscles.

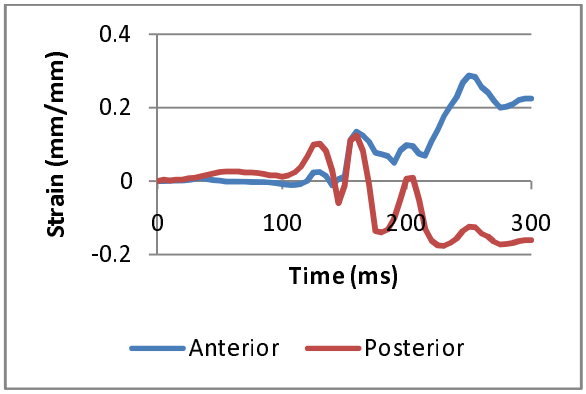


Figure 19. Predicted C5-C6 CL strain for a 10g impact with passive muscles.

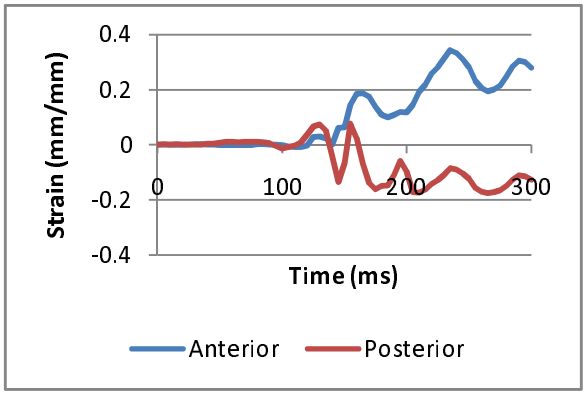


Figure 20. Predicted C5-C6 CL strain for a 12g impact with passive muscles.

Table 1.
Maximum CL strains in the model for different impact loads

	4g [2]	7g [1]	7g [2]	10g [1]	10g [2]	12g [1]
C2-C3	11.9	17.4	16.3	32.4	21.7	35.4
C3-C4	10.4	10.8	16.5	23.7	21.3	24.5
C4-C5	22.6	21.4	26.6	23.1	26.4	30.2
C5-C6	18.7	22.7	22.0	28.6	25.0	34.3
C6-C7	1.5	8.8	0.82	13.6	2.0	13.2
Maximum	22.6	22.7	26.6	32.4	26.4	35.4

1 Passive muscles
2 Active muscles

Table 2.
Model CL strains at various impact accelerations compared to published data (average (SD) in %)

	Model			Deng (1999a)		Panjabi et al. (1998)			Pearson et al. (2004)		
	4g	7g[1]	10g[1]	5g	6g	4.5g	6.5g	10.5g	3.5g	6.5g	8g
C2-C3	11.9	17.4	32.4	9.7	8.5	9.6 (13.7)	24.3 (26.3)	20.4 (5.3)	13.4 (9.3)	15.8 (13.5)	16.7 (6.3)
C3-C4	10.4	10.8	23.7	22.2	23.9	12.1 (4.0)	17.8 (13.6)	6.9 (0.3)	17.4 (15.2)	30.8 (25.1)	29.9 (17.8)
C4-C5	22.6	21.4	23.1	14.8	10.9	11.1 (15.8)	16.9 (10.5)	18.9 (13.8)	22.3 (20.6)	31.1 (22.5)	26.5 (18.7)
C5-C6	18.7	22.7	28.6	28.5	25.8	11.7 (5.1)	5.8 (7.5)	22.1 (15.1)	26.8 (17.9)	35.9 (21.9)	38.5 (24.6)
C6-C7	1.5	8.8	13.6			12.7 (N/A)	29.5 (25.7)	35.4 (N/A)	18.9 (14.2)	28.8 (20.0)	39.9 (26.3)

1 Passive muscles where used.

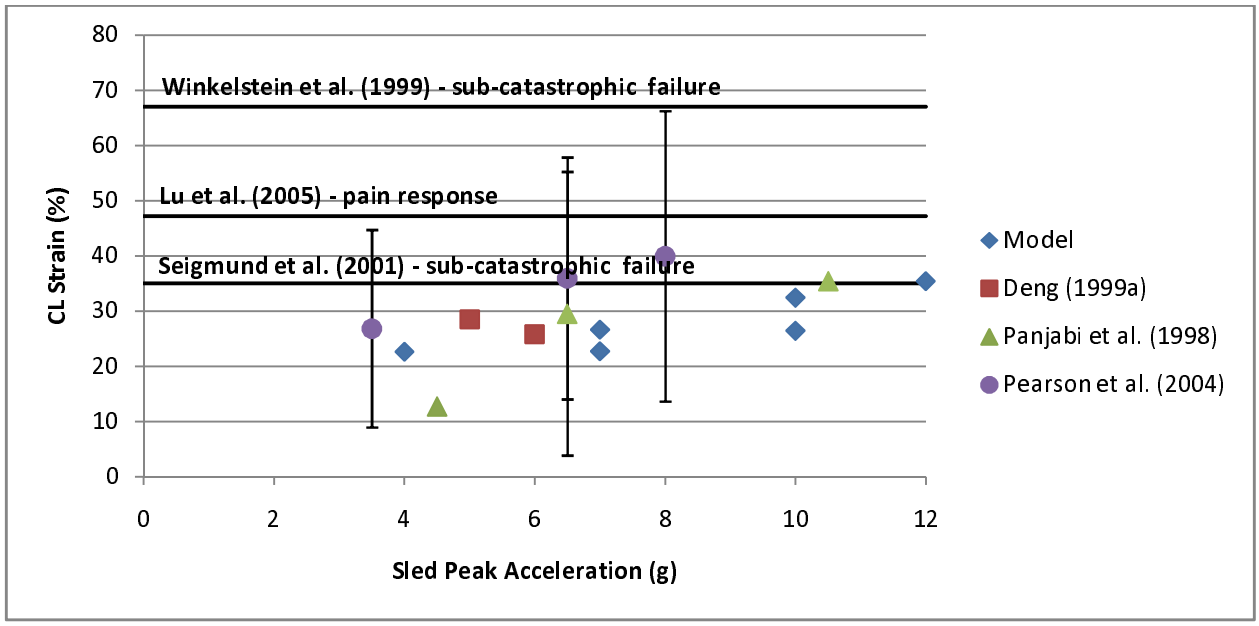


Figure 21. Maximum CL strain vs. peak sled acceleration for experimental results and the model. Standard deviation shown when available. Includes thresholds for injury.

DISCUSSION

Strain in the capsular ligament has been proposed as a possible source of neck pain for victims of rear impact and several clinical and biomechanical studies support this theory (Lord, et al. 1996, Barnsley, et al. 1995, Lu, et al. 2005, Lee, Davis, et al. 2004, Lee, Franklin, et al. 2006, Ivancic, et al. 2008, Quinn, et al. 2007, Panjabi, et al. 1998, Pearson, et al. 2004, Deng 1999a). A FE model of the cervical spine was used to measure capsular strains for rear impacts of 4, 7, 10g, and 12g. The results are unique from previous studies that have reported CL strain because the simulation included detailed facet joint modeling, active neck musculature, and full dynamic motion of the T1 in the sagittal plane.

The model was constructed accurately at the tissue level with the best available material properties, and was validated at the motion segment level and the whole cervical spine level for different frontal and rear impact scenarios. It should be emphasized that the model was not calibrated to any test conditions or data in order to preserve the accuracy at the tissue level, with the assumption that the CL strains measured during these simulations will be representative. The average CL strain was calculated by dividing joint distraction by ligament length and assumed a uniform strain state. Local CL strain could be predicted if shell elements were used for the ligament; however there is little literature available on local CL strain to justify this increase in model complexity. Another assumption made was that all the muscles contract at a given time, which was supported by EMG measurements of volunteers in rear impacts (Siegmund, Sanderson, et al. 2003, Ono, et al. 1997, Roberts, et al. 2002, Szabo and Welcher 1996). Further investigation into the effect of activate musculature on the CL strain during rear impact is required.

A possible limitation to the model was that the model was not calibrated to any of the impact tests. As a result, the model was in poor agreement with some of the experimental measures. The disagreement with the experimental impact data was more likely a limitation of the available tissue data used to develop the full spine model. In particular, some soft tissue characteristics such as viscoelasticity, nonlinearity, and anisotropy are not implemented in the model due to the lack of literature, and/or appropriate material models. The study and incorporation of these characteristics is a focus for model improvement in the future. Finally, it should be stated that the model is limited to the range of loads for which it has been validated.

The strains predicted in the CLs of the model have been compared to research performed by Deng (1999a), Panjabi et al. (1998), and Pearson et al. (2004) (Table 2). Deng used 6 cadavers in a series of 26 rear impacts, and measured the motion of each vertebra using a high speed X-ray to track implanted spheres and inferred a facet joint distraction. It should be noted that the strain results reported in Table 2 from Deng come from two tests, and have been scaled by a factor of Deng's initial gage length divided by CL lengths reported by Panjabi et al. (1998) for comparison to the current study. Panjabi et al (1998) used a specially designed spinal ligament transducer affixed across the facet joint to track joint distraction during rear impacts of four T1 to occipital cadaveric spines and divided that by anatomical ligament lengths to get strain Pearson et al. (2004). used a bench top sled with active muscle replication to impose rear impact loads on six occipital to T1 spine sections and measured vertebral motion by tracking marker flags using high speed cameras, and then inferred facet joint distraction that was divided by anatomical ligament lengths to get strain. The results in Table 2 show that for spinal levels C4-C5, and C5-C6 the predicted strains from the finite element model were within the quoted experimental range (Table 2). At the C6-C7 level the model predicted strains that were below the experimental range for all impact severities tested (Table 2). The model was in good agreement with the published data for the 10g case, but predicted a higher strain value at the C2-C3 level (Table 2).

In quasi-static spinal segment testing authors have found CL strains ranging from 11.6% to 17.8%, which are significantly lower than strains measured in dynamic tests (Table 2), demonstrating the importance of dynamic effects (Winkelstein, et al. 1999, Siegmund, Myers, et al. 2001). In an in-vivo study of goat facet joint distraction, Lu et al (2005) found what they hypothesized to be nociceptive (pain) receptors fire at a maximum principle strain of 47.2%. Testing of isolated facet joints from cadaver cervical spines have found sub-catastrophic damage to the CL at strains ranging from 35% to 66.8% in quasi-static testing, and 67% at 100mm/s (Winkelstein, et al. 1999, Siegmund, Myers, et al. 2001). At 10g the model predicted CL strain of 32.4% and 28.6% at the C2-C3 and C3-C4 levels respectively, which was just below the 35% threshold for sub-traumatic damage, and was well below the 47.2% threshold for pain. When the model was exposed to a 12g rear impact, the CL strain exceeded the 35% threshold reported by Seigmund et al. (2001). The model predicted CL strains have been

compared to experimental results, and published thresholds for injury (Figure 21). In a study of 28 instrumented real life accidents with 38 occupants, twenty occupants had short-term consequences at less than 10g, and two had long term consequences at 13g and 15g (Krafft, et al. 2000). In another study of 66 real life accidents, 13 of the 15 people that sustained neck injuries for longer than a month experienced a rear impact of greater than 9g (Krafft, et al. 2002). The 7g impact presented in this paper corresponds to an impact velocity of 7.5mph (Deng 1999a), and volunteer tests have been performed up to 6.8mph without mild symptoms, defined as lasting longer than 4days (McConnell, et al. 1995).

Future development of the cervical spine model will focus on improving the accuracy of the tissue models with the expectation that improved tissue models will improve the agreement between the full spine model and the experimental literature. This also includes a thorough investigation of the effect of active neck musculature on the response of the cervical spine in rear impact. The goal of this work is to better identify injury thresholds in rear impact scenarios, and to investigate out of position effects which has been suggested to increase strains (Winkelstein, et al. 1999).

CONCLUSIONS

The finite element cervical spine model used in this study, constructed from accurate geometry and the best available material properties, was previously validated at the segment level and for frontal impact scenarios. In this study, the model validated against volunteer and cadaver tests in rear impact scenarios and shown to be in good agreement. Capsular ligament strains predicted by the model approach thresholds for pain and sub-traumatic injury, but did not exceed them under 10 g rear impact loads which is consistent with the data in the literature. However, application of a 12g rear impact case did show higher strains that would be expected to result in pain or injury.

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