Out-of-Position Rear Impact Tissue-Level Investigation Using Detailed Finite Element Neck Model

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Objective: Whiplash injuries can occur in automotive crashes and may cause long-term health issues such as neck pain, headache, and visual and auditory disturbance. Evidence suggests that nonneutral head posture can significantly increase the potential for injury in a given impact scenario, but epidemiological and experimental data are limited and do not provide a quantitative assessment of the increased potential for injury. Although there have been some attempts to evaluate this important issue using finite element models, none to date have successfully addressed this complex problem.

Methods: An existing detailed finite element neck model was evaluated in nonneutral positions and limitations were identified, including musculature implementation and attachment, upper cervical spine kinematics in axial rotation, prediction of ligament failure, and the need for repositioning the model while incorporating initial tissue strains. The model was enhanced to address these issues and an iterative procedure was used to determine the upper cervical spine ligament laxities. The neck model was revalidated using neutral position impacts and compared to an out-of-position cadaver experiment in the literature. The effects of nonneutral position (axial head rotation) coupled with muscle activation were studied at varying impact levels.

Results: The laxities for the ligaments of the upper cervical spine were determined using 4 load cases and resulted in improved response and predicted failure loads relative to experimental data. The predicted head response from the model was similar to an experimental head-turned bench-top rear impact experiment. The parametric study identified specific ligaments with increased distractions due to an initial head-turned posture and the effect of active musculature leading to reduced ligament distractions.

Conclusions: The incorporation of ligament laxity in the upper cervical spine was essential to predict range of motion and traumatic response, particularly for repositioning of the neck model prior to impact. The results of this study identify a higher potential for injury in out-of-position rear collisions and identified at-risk locations based on ligament distractions. The model predicted higher potential for injury by as much as 50% based on ligament distraction for the out-of-position posture and reduced potential for injury with muscle activation. Importantly, this study demonstrated that the location of injury or pain depends on the initial occupant posture, so that both the location of injury and kinematic threshold may vary when considering common head positions while driving.

Keywords: biomechanics, finite elements, impact response, injury, modeling, neck injury, rear impact, whiplash

Introduction

The cervical spine is susceptible to injury from high loads or abnormal movement beyond the physiological range of motion; for example, from acceleration of the head in a vehicle crash scenario. Injuries are often described as traumatic, where catastrophic tissue failure occurs, and subtraumatic, where tissue damage may occur without catastrophic failure but resulting in pain response. The latter, corresponding to soft tissue strain or sprain in the neck, can lead to whiplash-associated disorders (WADs). Whiplash injuries are the most common injuries treated in emergency departments in U.S. hospitals (Quinlan et al. 2004) with incidence rates ranging from 28 to 834 per 100,000 population each year (Cassidy et al. 2000; Otremski et al. 1989) and recently estimated at 300 per 100,000 population in Western countries (Holm et al. 2009). The highest incidence rates have been observed in females between the ages of 15 and 24 (Quinlan et al. 2004; Spitzer et al. 1995) and occupant seat position and impact direction have been associated with WAD. Further, Jakobsson (2005) identified head-turned posture and increased head-to-headrest distance leading to significant increases in low-severity neck injury from rear impact. The annual cost of neck injuries is estimated at $4.5–29 billion in the United States and 5–10 billion Euros in Europe (Freeman 1997; Holm et al. 2009; Kleinberger 2000; Schmitt et al. 2014; Zuby and Lund 2010). WAD includes headache, shoulder pain, paresthesias, nausea, weakness, dysphagia, visual and auditory disturbance, and dizziness (Haldorsen et al. 2003; Klein et al. 2001; Norris and Watt 1983), which may persist for extended periods of time.
Potential mechanisms leading to pain response have been proposed, including ligament distraction, intervertebral disc distraction (Cronin 2014), muscle strain, and nerve root compression (Ivancic 2012; Nuckley et al. 2002). Neutral-position rear impact investigations have demonstrated that ligament strain, particularly in the facet joint ligaments, may exceed proposed thresholds, whereas disc strains and nerve root compression (intervertebral foramen height) were within typically range of motion values (Cronin 2014).

The source of WAD has been debated by many authors (Binder 2007); however, several anatomical sites have been proposed as potential sources of WAD, including the facet joints, dorsal root ganglia, spinal ligaments, intervertebral discs, vertebral arteries, and neck muscles (Sieg mund et al. 2009). Ligament distractions, in particular the capsular ligaments (CLs), have been proposed by several authors (Aprill and Bogduk 1992; Barnsley et al. 1994) as a potential source of WAD. Anatomic and histologic studies have identified mechanoreceptive and nociceptive nerve endings in ligaments (Cavanaugh 2000; Cavanaugh et al. 1989; Giles and Harvey 1987; Inami et al. 2001; Kallakuri et al. 2004; McLain 1994; Ohtori et al. 2003), and damage may lead to abnormal sensory signals and a decrease in neck mobility and proprioception (Panjabi 2006). Double-blind anesthetic block studies have identified 54–60% of whiplash patients with facet joint pain (Barnsley et al. 1995; Lord et al. 1996) and these findings have been supported by in vivo animal models (Lee et al. 2004; Lu et al. 2005). Experimental testing on isolated facet joint ligaments has identified subcatastrophic failure strains ranging from 35 ± 21% (Sieg mund et al. 2001) to 65 ± 74% (Winkelstein et al. 2000). Pearson et al. (2004) tested cervical spine cadaver models with muscle force replication in rear impact and concluded that capsular ligaments in the lower cervical spine, especially in C6–C7, are at risk of injury. Ivancic et al. (2004) used a similar setup and observed that anterior longitudinal ligaments experience high strains in the lower cervical spine with the highest strain at C6–C7 during an 8 g rear impact simulation. In frontal impact, Panjabi et al. (2004) observed significant increases in the strain at the supraspinous and interspinous ligaments and the ligamentum flavum and observed high strains of capsular ligaments for a 10 g impact. In out-of-position impact, Kaale et al. (2005) observed higher than normal deformation in the alar and transverse ligaments with magnetic resonance imaging (MRI) studies.

A study by Shugg et al. (2011) showed that driving tasks related to stopping increased the proportion of time spent in a nonneutral posture. Importantly, Sturzenegger et al. (1995) found that head posture during impact had a significant effect on the persistence of whiplash injury. In support of this theory, Kaale et al. (2005) used MRI signal fluctuations to suggest that patients in an out-of-position posture may have a higher potential for injury and identified the alar and transverse ligaments as potential locations of injury. This is further supported by the epidemiological findings of Jakobsson (2005), who identified increased risk of low-severity (Abbreviated Injury Scale 1) neck injuries in head-turned rear impacts.

Several approaches have been developed to evaluate neck injury risk, typically for use with anthropomorphic test devices. The most common method is the $N_q$ criterion, which was developed and primarily used for frontal neutral position impacts. This approach uses axial loads and bending moments in the sagittal plane to predict the potential for severe injury (Kleinberger et al. 1999). The neck injury criterion was developed for neutral-position rear impacts and evaluates injury based on the relative acceleration and velocity of the head and the first thoracic vertebra (Bosröm et al. 1996). Because these methods have not been validated for out-of-position scenarios and predict the likelihood of severe injury, they are limited in predicting the potential for WAD in out-of-position impact scenarios.

The benefits of predicting head response and the potential for injury at the tissue level have been demonstrated by several authors (Cronin 2014; Fice et al. 2011). Specifically, this approach allows for the investigation of potential injury locations and can assist in identifying injury mechanisms. However, studies to date have not been able to quantitatively compare the effect of position on tissue distraction and the potential for WAD relative to a neutral position. Although epidemiological studies have identified this effect, there are many confounding factors that make this information difficult to use with respect to safety system design. Experimental studies have measured kinematics and tissue distractions; however, they are limited due to possibly threshold differences between animal species and humans, and the lack of active musculature in postmortem human subjects (PMHS) and variability in the test data that may obscure important effects. There are also challenges with measuring ligament strains in the neck experimentally because the strain may vary throughout the ligament and the definition of length is somewhat arbitrary due to the large insertion area. Verified and validated detailed finite element (DFE) models can provide detailed kinetics and kinematics predictions for impact scenarios. A DFE is defined as a finite element model where relevant tissues are modeled with discrete elements and not with macroscopic approximations. For example, a detailed model of the intervertebral disc compared to a kinematic joint defined between 2 vertebral bodies. DFE can be used as a tool to investigate potential sources of injury, such as ligament distractions in the neck. Application of finite element models to investigate out-of-position has had limited success to date. Brolin and Halldin (2004) developed an FE model of the upper cervical spine requiring alteration of ligament stiffness to achieve good agreement with the literature. A subsequent parametric study identified the response of the upper cervical spine being sensitive to the capsular ligament properties but was limited by the calibrated material properties. Kallemyen et al. (2010) investigated a C2–C7 cervical spine FE model and identified the need to alter the available ligament properties to achieve published response in flexion, extension, lateral bending, and axial rotation. Zhang et al. (2006) developed a nonlinear FE model of the cervical spine and evaluated the model in flexion, axial rotation, and lateral bending. The authors concluded that the FE model offered potential for biomedical and injury studies; however, the lateral bending stiffness was high and the tissue properties were not published in detail, limiting applicability of the results. Storvik and Stemper (2011) studied the effect of axial head rotation on the facet joint capsule strains during rear impact using an FE model of the cervical spine validated in neutral position and not in axially rotated...
conditions, due to a lack of experimental data in literature. They demonstrated an increase in ligament strains from C3 through C7 due to an increase in axial rotation; however, the capsular ligament strains between C0–C2 and C7–T1 were not computed and the severity of impact was not found to have a significant effect on the capsular ligament strains. The model was not previously validated in range of motion or failure and demonstrated a larger than expected axial rotation between C1 and C2.

Analysis of out-of-position scenarios requires relevant data for range of motion and impact scenarios to evaluate the model performance. For this study, the upper cervical spine (C0 or skull, C1, C2) tests in the literature were used to evaluate 4 loading modes for range of motion and traumatic loading, including axial rotation (Goel et al. 1990), tension (Dibb et al. 2009), and flexion/extension (Nightingale et al. 2007). Whole cervical spine (C0–T1) testing was undertaken by Ivancic (Ivancic, Ito, et al. 2006; Ivancic, Panjabi, et al. 2006) to measure quasistatic response in axial rotation, extension, flexion, and lateral bending. The same test samples were then used for out-of-position dynamic impact testing. The experimental setup used cables and springs to replicate the passive musculature of the cervical spine and a 3.3 kg surrogate head rigidly attached to the occipital mount. Head–T1 rotations of 28.4° (5.3°) left axial, 3.5° (3.7°) flexion, and 17.9° (4.7°) left lateral bending were initially applied to the cervical spine, followed by T1 impact. The resulting relative motions of the vertebrae were measured and reported for the impact scenarios. Injury, defined by increased range of motion or changes in the neutral zone, was identified in the lower cervical spine, primarily at C5–C6 for 8 g impact scenarios, indicating a potential for injury in out-of-position impact scenarios.

The purpose of this study was to identify limitations of an existing DFE neck model to predict out-of-position response, determine a method to reposition the head while retaining the tissue stresses and strains, enhance and validate the model for out-of-position scenarios, and undertake an investigation to determine the effect of position on the tissue distractions and the potential for injury.

Finite Element Neck Model

The cervical spine finite element model used in this study represented a 50th percentile male, including 65,980 solid elements, 40,678 shell elements, and 4,412 axial elements (Fice et al. 2011; Panzer et al. 2011). The model was meshed using a commercial preprocessor (Hyperworks, Altair, Toronto, Ontario), solved using a commercial explicit finite element code (LS-Dyna version 971 R4.2.1, LSTC, Livermore CA) and included representations of the tissues relevant to kinematic response in the neck. Previously, the experimental data for individual ligaments of the upper cervical spine were limited and varied significantly between the few studies available (Panzer and Cronin 2009). The available data were implemented into the model but not validated outside of global head kinematics in frontal and rear impacts. In the current version of the model, the cervical spine ligaments were modeled using multiple nonlinear rate-dependent axial elements (DeWit 2012; DeWit and Cronin 2012) with a ligament-specific force-displacement response from experimental human tissue testing (Mattucci et al. 2012) for the lower cervical spine. The ligaments of the lower cervical spine included the anterior longitudinal ligament, posterior longitudinal ligament, ligamentum flavum, interspinous ligament, and capsular ligaments (Panzer and Cronin 2009). The upper cervical spine ligament properties were measured experimentally (Mattucci et al. 2013) and included the transverse ligament, anterior atlantooccipital membrane, posterior atlantooccipital membrane, capsular ligaments between skull–C1 and C1–C2, anterior atlantoaxial membrane, posterior atlantoaxial membrane, tectorial membrane, vertical cruciate, and apical and alar ligaments. The benefits of the approach used in the original model included direct application of the experimentally measured force-distraction data so that a ligament length and cross section, which vary within and between ligaments, was not required and incorporation of deformation rate effects in the tissue properties (DeWit and Cronin 2012). The passive and active muscle responses were modeled using 87 pairs of Hill-type axial elements for the 25 muscle pairs of the cervical spine (Fice 2010; Panzer 2006; Panzer et al. 2011) and the model has been validated extensively with experimental PMHS and volunteer data for neutral-position frontal and rear impact scenarios (Fice et al. 2011; Panzer et al. 2011). The intervertebral discs were modeled using shell elements with an orthotropic nonlinear material model to represent the annulus fibrosus fibers embedded in solid elements representing the ground substance and solid elements to represent the nucleus pulposus (Figure 1). The facet joint and upper cervical spine cartilage were modeled with solid elements and linear viscoelastic material properties. For this study, the applied loading was not expected to cause significant deformation or failure of the hard tissues, so the vertebrae and skull were modeled as rigid to reduce computation time. A side study with deformable vertebral bodies, meshed with solid elements representing the cancellous bone, and shell elements representing the cortical bone did not predict failure of the vertebrae bodies, verifying this assumption. At the segment level, the lower cervical spine had been previously validated in flexion, extension, lateral bending, and translation for range of motion and traumatic loading scenarios, demonstrating that the peak failure force or moment and the injury locations were in good agreement with the literature (DeWit and Cronin 2012; Panzer and Cronin 2009). The full cervical spine model was previously validated in tension and frontal (Fice et al. 2011) and rear impacts (Panzer et al. 2011), showing that the head kinematics and ligament strain response of the model were in good agreement with the literature (Fice et al. 2009).

It has been demonstrated that muscle activation is essential to model human volunteer tests (Brolin et al. 2005; Panzer 2011; Van der Horst 2002). Previous studies have measured electromyography signals approximately 60 to 79 ms following the initiation of an impact (Ono et al. 1997; Roberts et al. 2002; Siegmund et al. 2003b; Szabo and Welcher 1996). More specifically, the normalized peak signal for the paraspinal muscles (extensor) was measured to be 66 to 72% of the signal from the sternocleidomastoid (flexor), indicating that the flexor muscles activate with higher intensity (Braultz et al. 2000; Siegmund
Methods

The first phase of this study was to identify any limitations in the existing model for out-of-position impact scenarios, because the model had only previously been evaluated for neutral-position impacts in the sagittal plane. Initial simulations were undertaken and 3 areas for improvement were identified:

1. Existing repositioning methods apply displacement boundary conditions to obtain a nonneutral position and then restart the model with the tissue stresses and strains set to zero. However, out-of-position modeling for this study required that initial strains, particularly in the ligaments and disc, were retained. Several techniques were investigated, but many resulted in numerical instabilities and analysis termination due to the quasistatic nature of the repositioning followed by the dynamic impact event.

2. The upper cervical spine was not previously validated for range of motion and kinematics outside of the full neck frontal and rear impacts and was found to be overly stiff in axial rotation using the properties and test data available in the literature. The ligament properties of the upper cervical spine were improved using more recent experimental data (Mattucci 2011; Mattucci et al. 2013). Further, it was identified that ligament laxities, which are not measured in single-ligament experimental tests, were essential to model the axial rotation and flexion/extension range of motion response of the upper cervical spine. A method was developed to identify appropriate ligament laxities for this complex structure. Ligament laxity, specified as a dimension in millimeters, was represented in the FE model by allowing for initial distraction of the ligament without force response.

3. Ligament failure was not previously incorporated in the model. This was identified as important for high-severity loading and out-of-position scenarios where some ligaments may see larger distractions compared to a neutral posture. A ligament failure modeling technique that had been previously used in the validation at the segment levels (DeWit and Cronin 2012) was incorporated in the full cervical spine model for analysis of higher severity impacts that may result in significant ligament distractions and possibly failure.

The second phase of this study included model enhancement to address the above-noted limitations, followed by assessment and validation. A detailed investigation of upper cervical spine ligament laxity was undertaken using a functional spinal unit (FSU) extracted from the model (skull or C0, C1, C2) following implementation of the measured craniovertebral ligament properties from Mattucci et al. (2013) for a younger population (average age 45 years old) at strain rates relevant to automotive crash scenarios. The upper cervical spine ligaments in the model included the transverse ligament, anterior atlantooccipital membrane, posterior atlantooccipital membrane, CL between skull–C1 and C1–C2, anterior atlantoaxial membrane, posterior atlantoaxial membrane, and tectorial membrane/vertical cruciate/apical/alar ligament complex. In the current study, ligament laxity was defined as a value in millimeters corresponding to the distraction that occurred before a ligament exhibited a force response. This was achieved by offsetting the ligament force–displacement curves by an amount equal to the laxity for each ligament.

Ligament Laxity Study

To develop an understanding of the effect of ligament laxity on the range of motion, deformation to failure, and ultimate failure force, the upper cervical spine FSU model was investigated using 4 load cases: axial rotation (Goel et al. 1990), tension (Dibb et al. 2009), and flexion/extension (Nightingale et al. 2007). These experiments were modeled by fixing the skull and rotating or displacing the C2 vertebra. In a preliminary study (Shateri 2012), laxity values of −2 to 4 mm were considered to determine how each ligament affected the...
individual load cases. The results provided guidance for a second detailed investigation (see Figure B1, online supplement) where the ligament laxities were investigated in 0.5-mm increments and the failure force was required to be within the average plus or minus one standard deviation of the reported experimental data to be acceptable.

The full cervical spine was validated using 4 quasistatic rotation tests (66.8° in axial rotation, 54.6° in extension, 51.7° in flexion, and 30.4° in lateral bending) from cervical spine tests (C0–T1, average age 80.2 years; Ivancic, Panjabi, et al. 2006) that included a surrogate head but no actual or simulated musculature. In the experiments, the T1 vertebra was fixed while rotation was applied to the surrogate head in axial rotation, extension, flexion, or lateral bending and the relative rotations of the vertebrae were measured. The boundary conditions from the experiment were applied to the model with the skull properties set to those of the surrogate head (mass, 3.3 kg; inertia: sagittal inertia, 0.019; transverse inertia, 0.014; frontal inertia, 0.015 kg m²).

The ligament laxity investigation (Figure B1) began with axial rotation (range of motion and failure response; Goel et al. 1990). Axial rotation was investigated using both physiological range of motion and ultimate failure data. The model was then evaluated in extension and flexion (Nightingale et al. 2007) and finally in tension (Dibb et al. 2009). When a set of ligament laxities was determined for the upper cervical spine, the properties were implemented in the full cervical spine model and analyzed for comparison to the physiological range of motion data for the cervical spine (C0–T1; Ivancic, Panjabi, et al. 2006). The musculature was removed for this simulation because the experimental study did not include muscle force represented by the cables. At the full cervical spine level, the difference between the rotation values of the model and the average ± SD of the experimental data was calculated for all intervertebral bodies. The sum of these differences was divided by the total head rotation of the loading case. If the resultant value was less than 5%, the laxities were accepted; otherwise, the process was repeated again beginning at the segment level. The iterative analysis was conducted over several months to determine a set of ligament laxities. Optimization approaches, such as LS-OPT (LSTC), were investigated early in the study but were not pursued for 2 reasons. First, the method was very sensitive to the definition of limits on values (e.g., ligament laxity values) and could converge on answers that were not physically meaningful when not strictly confined. This was true for single modes of loading and this effect was exacerbated when simultaneously including the 4 modes of loading. Second, ultimate failure of the segment could occur by catastrophic failure of the tissues, resulting in a rapid release of stored energy and sudden large distractions, or by severe distraction of the joint, leading to large deformations with high forces or moments. These 2 different instability conditions were challenging to quantify in terms of requiring the optimization program to select appropriate values to determine the next iteration and required manual interaction to evaluate. For these reasons, automated optimization approaches were not pursued in this study but will be investigated further for future studies.

Dynamic Assessment Study

Following introduction of the upper cervical spine ligament laxity, the FE model was revalidated for neutral position rear impact scenarios to ensure that the model response was still within acceptable limits. The load cases included a 7 g rear impact cadaver study (Deng 1999) to validate the model with passive muscle for head kinematics, multiple 8 g rear impact cadaver tests (Ivancic et al. 2004; Panjabi et al. 2004; Pearson et al. 2004) for tissue-level validation, and a 4 g rear impact (Davidsson et al. 1998) for kinematic validation against volunteer experiments with active musculature. Detailed information on the load cases and data analysis is reported by Fice et al. (2011).

Following enhancement of the model and validation in neutral position, an out-of-position bench-top impact test of the whole cervical spine (Ivancic, Ito, et al. 2006; Ivancic, Panjabi, et al. 2006) was used to investigate the predicted intervertebral rotations for a head-turned impact scenario. To simulate the scenario, the mass and moment of inertia of the head were altered to those of the surrogate head used in the experiment. Initially, the model was modified to incorporate the experimental spring/cable musculature, but sufficient detail was not available to properly implement this approach, so the original model musculature was used with no muscle activity (i.e., passive neck musculature). The average head rotation from the experiment was applied to the FE model using displacement controlled rotation followed by an 8 g rear impact applied to the T1. The head kinematics and the intervertebral rotations from the FE model were compared to the experimental data. The third phase of this study was to apply the enhanced DFE to investigate the effect of axial head rotation and muscle activation in different impact severities. The model was positioned with an axial head rotation of 42.5°, which was the average right head rotation reported by Shugg et al. (2011), representative of the rotations during stops and lane changes when driving. The head rotation was applied to the FE model by axially rotating the head about the occipital condyle while it was free to move in all other directions. After the rotation, accelerations of 7, 12, or 16 g (Fice et al. 2011) were applied to the T1 vertebra to simulate rear impacts. Both passive and active musculature scenarios were used to investigate the effect of position and muscle activity on tissue distractions and the potential for pain response, resulting in 12 impact scenarios.

Results

Ligament Laxity Study

The effect of varying individual ligament laxity from −2 to 4 mm was investigated to provide an initial frame of reference for the optimization study. It was found that pretension (e.g., −2 mm laxity) resulted in nonphysical response of the FSU (uncharacteristic stiffness and failures) and so results for laxity from 0 to 4 mm were reported as primary (if the response changed by more than 10%) and secondary (if the response change was between 5 and 10% of the target value). The axial rotation mode of loading was most sensitive to ligament
laxities, with 9 of the ligaments having a primary effect (Table B1, see online supplement) on the response. Extension was the second most sensitive load case, with 5 ligaments having a primary effect.

The laxities for the ligaments of the upper cervical spine (skull or C0, C1, and C2) were determined using an iterative method (Figure B1) with 4 different load cases: axial rotation, flexion, extension, and tension. Determination of the final ligament laxities required several iterations, each including subiterations to identify appropriate limits or boundaries for the laxity values. The iterative procedure was terminated when the available responses were within ±1 standard deviation of the average values, as reported in the literature. The resulting ligament laxities (Table B2, see online supplement) improved the torque and ratio of C01/C12 rotation at the segment level during physiological range of motion in axial rotation (Figure 2). The failure load predicted by the FE segment model was within the standard deviation of reported values in the literature in axial rotation, tension, flexion, extension (Figure B2, see online supplement). The predicted deformation to failure was within the standard deviation in most cases.

**Dynamic Assessment Study**

Following determination of the craniovertebral ligament laxities, the model was evaluated for rear impact scenarios with the head and neck initially in a neutral position. This revalidation procedure was undertaken to ensure that the introduced laxities did not significantly alter the model behavior in neutral position, compared to the original model response and experimental data. The global kinematics and the local tissue response (Appendix A, see online supplement) were in good agreement with the previous work by Fice et al. (2011) when compared to PMHS rear impact test head kinematics (Deng 1999), vertebral body relative rotation, maximum ligament strains, and a lower severity rear impact scenario (Davidson et al. 1998). This was important to verify because the previous model had undergone significant development and validation using a wide range of loading cases. Importantly, the addition of laxity to the upper cervical spine significantly improved axial rotation response, particularly for range of motion response, but did not significantly affect the global head kinematics for the impact cases considered.

Evaluation of out-of-position scenarios began by comparing the model to experimental test data from a bench-top 8 g rear impact test of the whole cervical spine (Ivancic, Ito, et al. 2006; Ivancic, Panjabi, et al. 2006) with a surrogate head mass. Muscle force replication was incorporated using springs and cables, which could not be reproduced in the model due to limited information in the study, and was represented by the passive musculature as implemented in the model. This was reasonable considering the purpose of the springs and cables was to reproduce the effect of the musculature. The predicted model head kinematics (Figure B4, see online supplement) and the intervertebral rotations (Figure B5, see online supplement) were similar to the reported experimental data in axial rotation and flexion/extension. The behavior of the FE model deviated from the experimental results in lateral bending and flexion after approximately 150 ms. The predicted intervertebral rotation generally followed the experiment in extension and axial rotation, when compared to the experimental results (Ivanic, Panjabi, et al. 2006) with deviations occurring at the C5/C6 level in the model.

**Parametric Study**

The out-of-position study investigated the effect axial head rotation and muscle activation on ligament distraction in rear impact scenarios. The model predicted a higher potential for ligament injury, based on ligament distraction, at higher impact levels (Figure 3), as expected. This increase was seen in the majority of the ligaments and was amplified in the out-of-position posture compared to neutral posture. Muscle activation was shown to reduce the potential for injury in both neutral and nonneutral postures (Figure 4). During out-of-position rear impacts, the model predicted higher potential for injury at most impact levels with both passive and active
Fig. 3. Effect of impact severity on ligament strains for capsular ligaments: (A) neutral position and (B) 42.5° axial rotation.

muscle types (Figure C1, see online supplement). This increase was mostly notable in the ligaments of the upper cervical spine.

Discussion and Limitations

Ligament Laxity Study

Detailed finite element neck models have not previously been successfully evaluated in axial rotation for the purpose of predicting kinematics and tissue-level response in impact. Initial studies with an existing detailed finite element neck model demonstrated high axial rotation stiffness, particularly in the physiological range of motion. This is a challenging issue to address because existing experimental studies provide force–displacement data for individual ligaments and identify the initial condition with laxity removed by preconditioning the sample. The significant range of motion and ultimate

Fig. 4. Effect of muscle activation on ligament strains for (A) capsular ligaments: neutral position; (B) capsular ligaments: 42.5° axial rotation; (C) upper cervical spine ligaments: neutral position; and (D) upper cervical spine ligaments: 42.5° axial rotation. Suffix A represents active muscles and P represents passive muscles. Subcatastrophic values from Siegmund et al. (2001); failure corridor from Mattucci et al. (2012, 2013).
strength of the upper cervical spine is controlled by 15 ligaments that make selection and optimization of ligament laxities challenging. More specifically, several nonunique solutions were identified in initial analyses, highlighting the need for a methodical approach to first approximate the required laxity values and understand the contribution of each ligament to the response using the limited data in the literature, followed by a detailed study to estimate the ligament laxities.

The addition of laxity to the ligaments in an isolated upper cervical spine model provided a good understanding of how laxities affect the loads during flexion, extension, tension, and axial rotation. The laxities of the anterior ligaments affected failure loads and deformations in extension, tension, and axial rotation. The posterior ligament laxities primarily affected flexion response, whereas the capsular ligament laxities affected all loading conditions and axial rotation in particular. The alar ligament largely affected extension and axial rotation. The laxity in many of the upper cervical spine ligaments affected the torque and axial rotation in physiological range of motion (Table B1). The initial study provided a first estimate of ligament laxity for the subsequent iterative process. This preliminary study also highlighted the importance of each ligament in various loading conditions and was used to determine the order of optimization (Figure B1) based on the response sensitivity to the different modes of loading. Optimization software (LS-OPT, LSTC) was investigated but found to be sensitive to the definition of solution limits, and tissue failure and instability in the functional spinal unit were difficult to quantify in an automated method. For these reasons, automated optimization approaches were not pursued in this study but will be investigated further for future studies.

The resulting laxities for the upper cervical spine ligaments (Table B2) improved the axial rotation behavior for the physiological range of motion, which was essential to ensure proper response for repositioning (axial rotation) of the neck model prior to impact. The predicted torque for range of motion was reduced by approximately a factor of 5 compared to the previous model without laxity, and fell within the standard deviation reported in the literature (Goel et al. 1990). The ratio of C01/C12 rotation at the segment level was considered essential because the largest axial rotation in the neck occurs between the atlas and the axis vertebrae. After the addition of laxities, the atlas–axis axial rotation ratio of the neck was closer to the 5–20% ratio (28%) reported in the literature (Dugailly et al. 2010; Dvorak et al. 1987; Ishii et al. 2004; Ivancic, Panjabi, et al. 2006; Figure B2). The failure loads were within one standard deviation of the average experimental results in all loading cases.

In the full cervical spine model, the laxities improved the intervertebral axial rotations, as reported in the literature (Ivancic, Panjabi, et al. 2006). Similar to the segment level, the C1–C2 to total head rotation was vastly improved for axial torque and rotation. This was primarily due to the laxities added to the capsular ligaments between the atlas and the axis, which is also supported in the literature (Broslin et al. 2004). The upper cervical spine intervertebral rotation was improved in extension mostly due to the added laxities in the anterior atlanto-occipital membrane and anterior atlanto-axial membrane ligaments. The improvement of the upper cervical spine behavior in flexion was largely due to the laxities in the posterior atlanto-occipital membrane, posterior atlanto-axial membrane, and interspinous ligaments between the atlas and the axis. In lateral rotation, the intervertebral rotations were in good agreement with the literature before adding laxities. However, the addition of laxity only resulted in small changes, and the values with upper cervical spine laxity were still in good agreement with the literature.

The relative rotation between the vertebrae for the full cervical spine was in good agreement with the literature (Ivancic, Panjabi, et al. 2006; Figure B3, see online supplement). The introduced laxities for the upper cervical spine ligaments dramatically improved the axial rotation ratio of the atlas and the axis within the full cervical spine model. In extension and flexion, the addition of laxities improved the intervertebral rotation of the upper cervical spine. Overall, the introduced laxities provided responses that were within one standard deviation of the average response and improved all responses by moving them closer to the average values.

**Dynamic Assessment Study**

In rear impact scenarios, the DFE model showed good agreement with the literature in neutral position collisions (Appendix A) with no significant change in the time history shape and less than a 5% change in peak response for the cases considered. Interestingly, the laxities introduced in the craniovertebral ligaments did not affect head kinematics for an initial neutral position. This may explain why ligament laxity has not previously been identified as important for DFE cervical spine models that have primarily been evaluated in a neutral position. In out-of-position scenarios, the model was compared to an experimental cadaver model that used muscle force replication (cables and springs) for passive muscle. A limitation of this case was that the simulated musculature (cables) could not be properly modeled due to lack of information on precise locations of the cable attachments and the complex sliding behavior of the cables through the pulleys and wire loops. The model was evaluated using the integrated passive musculature and demonstrated general agreement with the experimental results (Ivancic, Ito, et al. 2006; Ivancic, Panjabi, et al. 2006) in extension and axial rotation. The model head was identified as moving to the neutral position earlier in flexion and lateral bending compared to the cadaver test. This behavior may be caused by the different muscle implementation between the physical and DFE models.

In the current study, the model was validated based on the average ±1 SD limits reported in the literature. One limitation to this approach is that the values reported in the literature and the variability within the data may differ due to experimental methods and factors such as the number and age of samples. Importantly, however, the model predicted response within the one standard deviation of the average experimental data for a wide range of load scenarios (range of motion to traumatic levels) and modes of loading.

**Parametric Study**

The neck model was enhanced for out-of-position modeling through the implementation of ligament laxity in the upper
cervical spine, methods for repositioning the head, and comparison of the model to experiments in neutral position and head-turned rear impacts. Subsequently, a study was undertaken to evaluate the effects of head-turned impact severity and muscle activity on response at the tissue level. This study predicted that the potential for ligament injury increased at higher impact severities, as expected, because higher T1 accelerations generally lead to larger head rotations and ligament distractions, although other modeling efforts have not demonstrated this effect (e.g., Storvik and Stemper 2011). Most of the ligaments in the cervical spine were affected by impact severity. One exception was the posterior ligament in the lower cervical spine, which did not experience excessive strains due to the motion being primarily extension in that region of the neck. The increase in potential for injury due to impact severity was observed for neutral and nonneutral postures. In nonneutral postures, the upper cervical spine capsular ligament strains were higher. This is an important finding because capsular ligament distraction is a potential source of pain in the cervical spine (Siegmund et al. 2009).

The parametric study included scenarios with active muscle compared to the baseline passive muscle response. In general, the ligament strains in the neck were reduced with active musculature. Muscle activation provided additional resistance, reducing the neck motion and ligament strains. A similar effect has been observed in neutral position impacts for a DFE numerical model (Fice et al. 2011) and a multibody neck model (Van Der Horst et al. 2002). The muscle activation and reduced distraction had a significant effect on protecting the ligaments of the upper cervical spine. This was also reported previously for frontal impact simulations (Brolin et al. 2005). Muscle activation had a larger effect on reducing strains of the capsular ligaments and upper cervical spine ligaments such as the alar, apical, and inferior crux ligaments during out-of-position impacts compared to neutral-position impacts.

A higher potential for ligament injuries was identified for out-of-position rear impact scenarios compared to the same impact scenarios with an initial neutral position. There was more than a 50% increase in the upper cervical spine and the C6–C7 capsular ligaments, a potential source of pain, during a 16 g out-of-position rear impact. The alar ligament was identified as having increased distraction in out-of-position scenarios. Kaale et al. (2005) identified this ligament as a potential pain source for WAD. The model predicted over 15% increase in the alar and the apical ligament strains during a 16 g out-of-position rear impact, compared to the same impact in a neutral position. In the middle cervical spine region, the C2–C3 posterior longitudinal ligament and the ligamentum flavum were predicted to exhibit a 30% increase in strain and showed potential for failure with initial axial rotation of the head followed by a 16 g rear impact. The 30% increase in anterior longitudinal ligament distraction predicted by the model at the C6–C7 level was significant and could exceed accepted distraction thresholds for pain response. The model also predicted increases in strain for other ligaments, but these were not identified as exceeding existing tolerance thresholds.

Previous research has highlighted the importance of neck position on the potential for injury, but to date this has not been successfully investigated using finite element models due to the complexity and large range of motion of the upper cervical spine and the need for active musculature.

A detailed finite element model incorporating active musculature, previously validated for frontal and rear impacts, was investigated for out-of-position impact scenarios. The primary limitation was the need for laxity in the upper cervical spine. An iterative method was undertaken to determine the ligament laxities using experimental studies for the isolated upper cervical spine and whole neck, providing response in axial rotation, tension, flexion, and extension that met the reported range of motion and traumatic loading data. The full neck model was revalidated for rear impact. It was noted that the upper cervical spine ligament laxities did not significantly affect head kinematics in neutral position frontal or rear impact. A head-turned rear impact scenario reported in the literature was modeled and the results were in general agreement. Differences in lateral bending were attributed to the use of the integrated passive musculature in the model, whereas the experimental used a complex cable, spring, and pulley muscle force replication system, which could not be reproduced in the model.

The presence of active musculature was found to reduce ligament strains for initially neutral and head-turned postures. Axial rotation of the head in an out-of-position scenario resulted in increased ligament distractions and increased potential for pain response, particularly in the alar and apical ligaments. Importantly, this study demonstrated that the location of injury or pain depends on the initial occupant posture, so that both the location of injury and kinematic threshold (i.e., impact severity to cause pain) may change when considering common head positions while driving.

The current model demonstrates good prediction capabilities at the segment and full neck levels for multiple modes of loading. In particular, the model has been enhanced to predict axial rotation response and head-rotated out-of-position impact response. Interestingly, incorporation of the upper cervical spine ligament laxity did not significantly change the head kinematics in a neutral posture, indicating that this is something that may not be identified in models with an initial neutral position. Further research on muscle activation strategies and the potential for muscle injury as a pain source are needed.

The results of this study identify a higher potential for injury in out-of-position rear collisions and identified at-risk locations based on ligament distractions. This development provides a basis for future work in more complicated loading scenarios including vehicle rollover and small offset frontal crash.

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Supplemental Materials

Supplemental data for this article can be accessed on the publisher’s website.

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