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Numerical analysis of compressive neck injuries due to vertical loading applied through the head

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Abstract:

Background: The vehicle injury priority rating data indicated that neck injuries became the fifth most important injury category (after head, face, chest and abdomen). During rollovers the centrifugal force tends to maintain a belted occupant erect with his/her head upward and outboard, and an impact to the occupant's head would come from the upper and lateral sides of the vehicle. Up to now, knowledge on the mechanism causing this kind neck injury is still rather limited.

Objective: The purpose of this paper is to summarize a numerical analysis of compressive neck injuries due to vertical loading applied through the head.

Method and Material: Our own validated, custom-made head-neck complex FE model was used to simulate those cadaver compression tests, which was integrated 50th percentile human head-neck model with various ligaments, joint capsule and muscles connecting the head and the neck.

Results: Comparing all the cases, in general C2-C3 facets were compressed most in both local x-axis (along the facet joint surface) and z-axis (normal to the facet joint surface) more than other facets, and severe disc compression can be seen in C5-C6 and C7-T1 discs.

Conclusion: The role of compression force with more sliding in cervical facet surface may be an important causal mechanism leading to facet failure in compressive neck injuries.

Keywords: Neck injury, Facet capsule, Rollover crashes, Numerical analysis

1 Introduction

The vehicle injury priority rating data indicated that neck injuries became the fifth most important injury category (after head, face, chest and abdomen). Previous experimental (Bahling et al, 1990) and numerical (Hu, 2008) studies have shown that during rollovers the centrifugal force tends to maintain a belted occupant erect with his/her head upward and outboard, and an impact to the occupant's head would come from the upper and lateral sides of the vehicle. Therefore, a combined lateral bending, compression, and flexion/extension could be a very common neck loading mode during rollovers. Up to now, knowledge on the mechanism causing this kind neck injury is still rather limited.

For this reason, this paper summarizes a numerical analysis of compressive neck injuries due to vertical loading applied through the head.

2 Method and Material

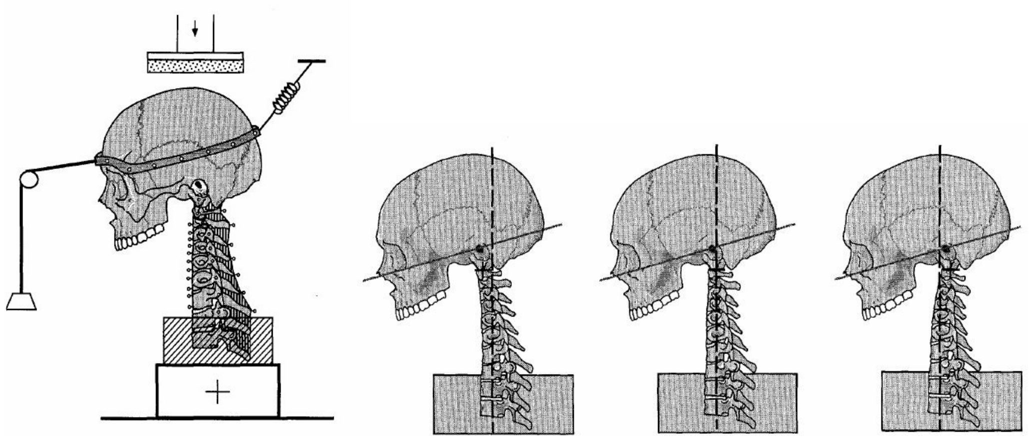
2.1 Cervical compression test simulations

Pintar et al in 1995 performed dynamic compression tests on a total of 20 human cadaver head-neck complexes (Figure 1). Our own validated, custom-made head-neck complex FE model was used to simulate those cadaver compression tests, which was integrated 50th percentile human head-neck model with various ligaments, joint capsule and muscles connecting the head and the neck.

To mimic the test condition, the FE neck model was fixed at the T1 inferior surface using the casting model as shown in Figure 2. The integrated neck-head model developed in neural position was rearranged to the stiff axis (Figure 2) by leveling vertebrae axis in alignment with the location of the occipital condyle. To simulate flexion and extension positions of the neck axis as described in experimental setup, the FE model was modified by moving the head 5 mm in the positive and negative x directions. The muscles were removed from the FE model. The casting was fixed in all directions at the inferior surface. A constant velocity loading was given to the impactor coated with a layer of ensolite. Only the average velocity of 5.25 m/s was used because the paper by Pintar et al. did not report exact velocities for each case. The additional velocities of 3.25 m/s and 7.25 m/s were simulated as well. The impactor was given at above velocities to a distance of 24 mm from the point of contact.

2.2 Facet capsule strain

Having validate force-deflection responses of the FE neck model against results from cervical axial compression tests performed by Pintar et al. 1995, the compression and facet strain from all three loading conditions (straight/compression, flexion and extension) were calculated for three impact velocities (3.25 m/s, 5.25 m/s, 7.25 m/s). To calculate facet stretch, the local co-ordinate system (body fixed frame) within each adjacent vertebra next to facet surfaces was defined and any movement in the +X and +Z axes was assumed to be tension and in -X and -Z axes to be compression. The facet stretch under cervical axial compression was calculated by measuring the movement of the n^{th} inferior facet cartilage surface with respect to $n+1^{th}$ superior facet cartilage surface in both local x-axis (along the facet joint surface) and z-axis (normal to the facet joint surface)



(a) **(b) flexion** **stiff** **extension**

Figure 1. (a) Experimental setup showing the positioning of the cadaver and loading applied

(b) Experimental positions (flexion, stiff and extension) inducing flexion, compression and extension type injuries.

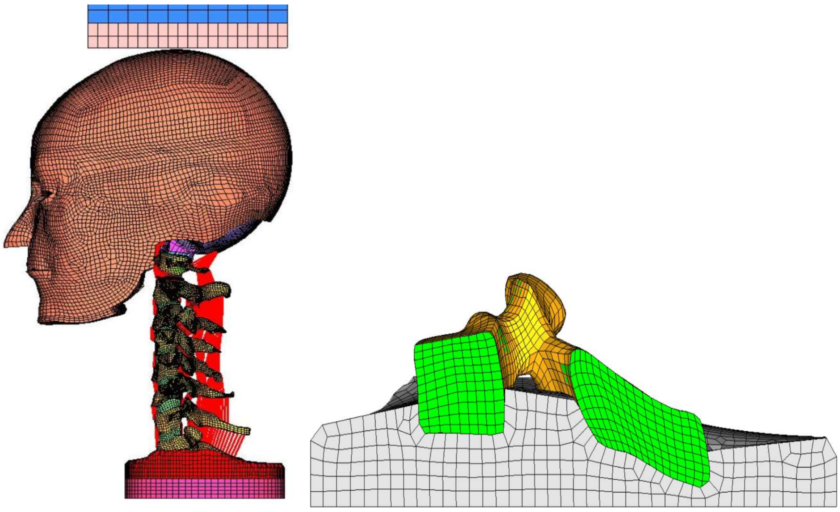


Figure 2. (a) FE model setup (b) T1 fixed to the casting

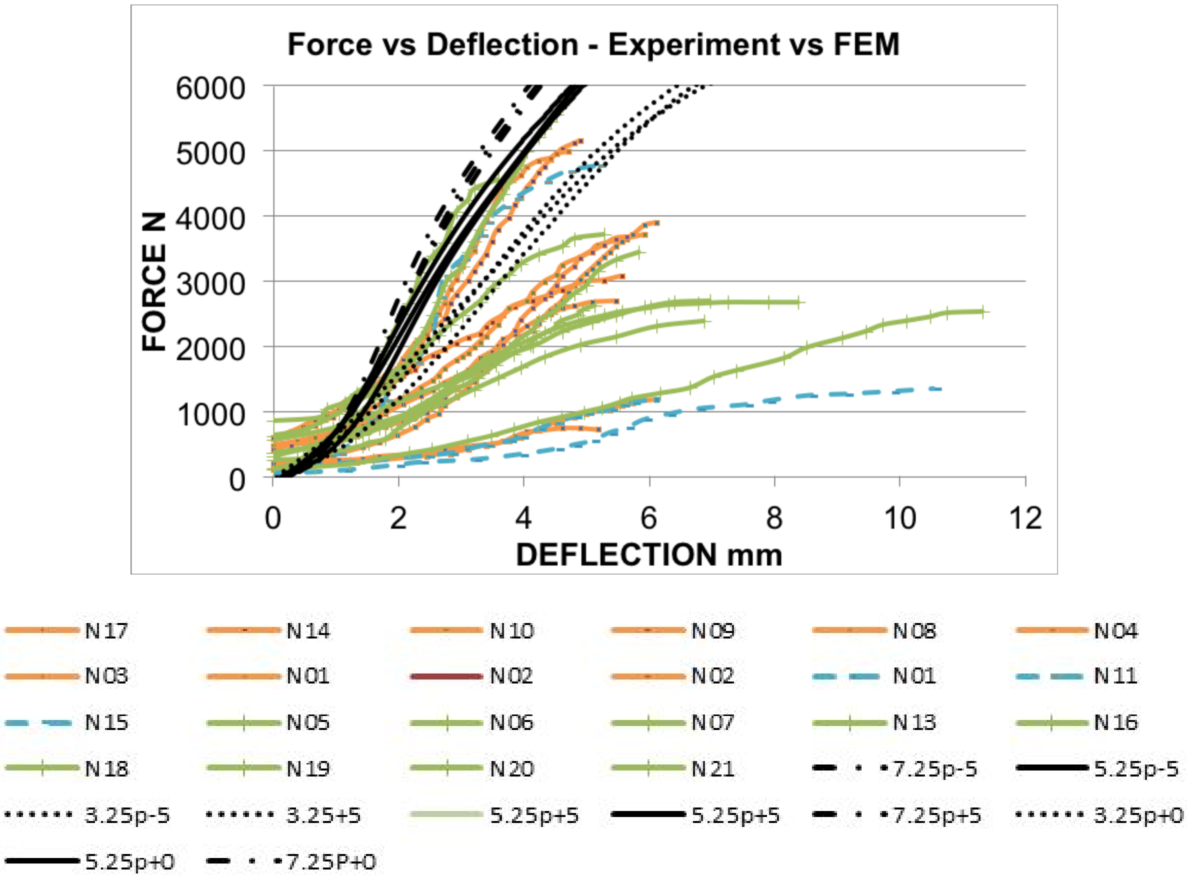
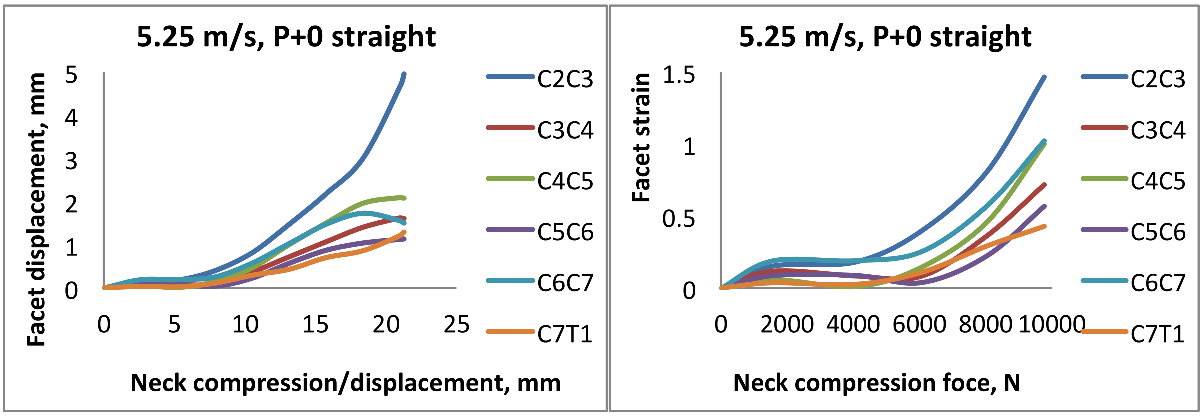


Figure 3. Model predicted neck force-deflection vs experiment results by Pintar et al. 1995



(a) (b)

Figure 4. Facet resultant displacement/strain as functions of neck compression (a) and compression force (b) with more sliding occurs in C2-C3 facet in straight neck position from 5.25 m/s impact (Pintar et al. 1995).

3 Results and Discussion

3.1 Compression force vs compression deflection

The cervical axial compression experiments done by The Medical College of Wisconsin are very different from the drop test. Here the impactor is maintained at a constant velocity and the axis of the spine is straightened or a "stiff axis" is maintained by removing the cervical lordosis. It produces a direct loading to the spine and gives very little space for the viscoelastic components to dissipate energy during the loading.

The FE model was rearranged to the stiff axis by bringing the vertebrae to the stiff axis by leveling it to the location of the occipital condyle. The intervertebral discs and facet capsules were created again for this loading. The muscles were removed from the FE model. A constant velocity of 5.25 m/s was used because the paper did not report exact velocities for each case. To add information for model sensitivity additional velocities of 3.25 m/s and 7.25 m/s were simulated. The head was moved 5 mm in the anterior and posterior direction to simulate flexion and extension injuries as reported in the paper.

The FE model was simulated to make the impactor move down to a distance of 24 mm from the point of impact to ensure the stability of the FE model (Figure 3). The FE model force response for 3.25 m/s and 5.25 m/s was in the range of force versus deformation plots provided by the researchers. The higher velocity impact gave a high force response as the impactor travelled down.

Since there is not much information given about the velocity of impact, we cannot give a proper conclusion about the reasons behind a higher force peak at a distance close to 12 mm from the point of impact.

All the FE simulations resulted in a compression type injury for 7.25 m/s cases. The 5.25 m/s cases had a buckling response at the C2-C3 level for the flexion and extension cases. The 3.25 m/s case had a similar injury pattern. No further study was made regarding the model sensitivity in this validation due to lack of information from the paper.

Vertebral body compression was noted at the C7 vertebra in all the cases and very small deformation was observed in the C5 and C6 level. Mid column fractures reported by Pintar et al. can be seen in the simulation. These experiments did not report any facet joint injury. The major areas of injury were vertebrae, ALL and PL. In the FE model, no vertebral body compression was seen except for 7.25 m/s case were a minor C7 vertebral body compression occurred. At higher velocities, the skull fractured and was still able to contribute a higher force on the C7 vertebral body. This makes the C6-C7 and C7-T1 region a potential injury site.

Comparing all the cases, severe disc compression can be seen in C5-C6 and C7-T1 discs. In the case of flexion setup where the head was moved 5 mm from the stiff axis, the lowest velocity case produced bending in the cervical spine whereas the other two cases produced a compression type injury with minor bending at the C4-C5 level. The C4-C5 disc experiences a partial shear loading due to the shape of the C5 vertebral body. So this becomes a typical rotation center during the flexion setup. This bending produced a severe posterior C2-C3 disc compression. In the extension type setup C2-C3 disc compressed more than the other discs. In all the cases the C3-C4 and C4-C5 disc was less vulnerable to injury. There was no evidence of spinous process fractures in all the cases.

3.2 Axial compression vs Facet strain

At 5.25 m/s impact for example, in case while the neck was in straight position, the compressive force was over 7000 N. The C3-C4 and C4-C5 facets experienced high compression in the X axis. C7-T1 facet experienced tension along X axis. The facetssustained compression in both X and Z axes. The C7-T1 facets experienced a negligible facet movement in both the axes. Same trend was observed in the flexion position case except C2-C3 facetswhich showed higher facet compression in Z close to 1.5 mm.For extension position case, C7-T1 facets tensed in X axis and the C4-C5 facets sustained a greater facet compression of 1.5 mm in the X axis than others. The facet movement in Z axis was mostly compression with C6-C7 facet compressed more than the previous facets. For flexion and extension cases, the compression force reached over 7800 N.

For all the six cases, in general C2-C3 facets were compressed most in both axes more than other facets. The compression of the vertebral body transformed the force to the C2-C3 facets which were in an oblique angle and therefore were forced to move along the face surface in X and Z. This indicated that increases in compression may lead to increases in facet strain subjected to vertical loadings (Figure 4). The C7-T1 facets experienced a compression in the earlier stage of the loading followed by extension as the force dropped down due to the relative movement of the C7-T1 vertebrae occurred at the end of the simulation.

4 Conclusions

The numerical analysis which was based on cadaveric cervical axial compression studies by Medical College of Wisconsin can be classified as severe impacts to the spine occurring during rollover car crashes. In that test, the spine stiffens out before impact and when the roof contacts the skull, the loading is directly applied to the spine. Pintar et al. have brought the “stiff axis” position to directly load the cervical spine components.

Comparing all the cases, in general C2-C3 facets were compressed most in both local x-axis (along the facet joint surface) and z-axis (normal to the facet joint surface) more than other facets, and severe disc compression can be seen in C5 C6 and C7 T1 discs. Of note, the positive correlation between compression and cervical facet capsule strain suggested important mechanisms occurred during this crown impact. The role of compression force with more sliding in cervical facet surface may be an important causal mechanism leading to facet failure in compressive neck injuries. More simulations of facet strain from cadaver tests are needed to confirm the hypothesis.

Acknowledgements

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