ISSN : 0974 - 7435 Volume 10 Issue 23

2014

BioTechnology
An Indian Journal

FULL PAPER

BTAIJ, 10(23), 2014 [14534-14539]

Numerical analysis of compressive neck injuries produced by the crown impact to the head

H.B.Chen[1,2*], L.Y.Zhang[2], J.H.Zhou[1], S.Tang[1], L.He[1], J.Zhang[1,3]
[1]State Key Laboratory of Trauma, Burns, and Combined Injuries, Institute of Surgery Research, Daping Hospital, Third Military Medical University, Chongqing 400042, (CHINA)
[2]Bioengineering Center, Wayne State University, 818 W. Hancock, Detroit, MI 48201, (USA)
[3]College of Optoelectronic Engineering, Chongqing University, Chongqing 400044, (CHINA)
E-mail: hoychen_mail@126.com

ABSTRACT

Traffic trauma data indicated that neck injuries became the fifth most important injury category (after head, face, chest and abdomen). A vertical compression from the crown impact to the head could be a very common neck loading mode during rollovers. The purpose of this paper is to summarize a numerical analysis of compressive neck injuries produced by the crown impact to the head. In this paper, a validated head-neck complex FE model was used to simulate those cadaver compression tests, and the compression and facet strain from all three loading conditions (flexion, stiff and extension) were calculated for three impact velocities (3.25 m/s, 5.25 m/s, 7.25 m/s). Comparing all the cases, it was suggested that: ① A high compressive load can be transmitted from the head to the neck only if the cervical spine was straight. Especially, at higher crown-impact velocities the skull fractured and was still able to contribute a higher force on the C7 vertebral body, making the C6-C7 and C7-T1 region a potential injury site. ② When the neck is in the flexed or extended position initially, the C4-C5 intervertebral disc becomes a typical rotation centre so that C3-C4 or C4-C5 disc is less vulnerable to injury. ③ The C2-C3 facet is likely to be compressed most in both local x-axis (along the facet joint surface) and z-axis (normal to the facet joint surface) and 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 joint; Rollover crashes; Numerical analysis; Compression.

INTRODUCTION

The human neck has a very complex anatomical structure, comprising seven bony segments, typically referred to as C1 to C7, with cartilaginous discs between adjacent vertebral bodies. The neck supports the weight of the head and protects the nerves that conduct sensory and motor information from the brain down to the rest of the body. Traffic trauma data indicated that neck injuries became the fifth most important injury category (after head, face, chest and abdomen)[1-3]. Previous experimental[4] and numerical[5] studies have revealed 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 vertical compression induced by the crown impact to the head 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 summarized a numerical analysis of compressive neck injuries produced by the crown impact to the head.

METHOD AND MATERIAL

Cervical compression test simulations

Pintar et al in 1995[6] performed dynamic compression tests on a total of 20 human cadaver head-neck complexes (Figure 1a-1b). A 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.

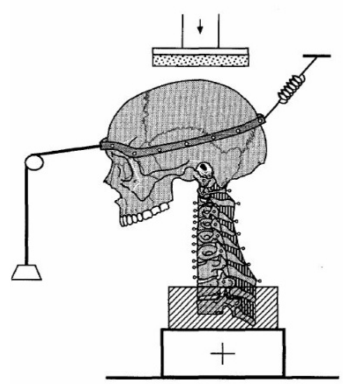

Figure 1a : Experiment setup showing the positioning of cadavers.

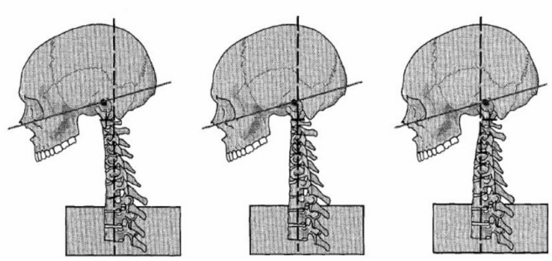

Figure 1b : Experimental positions inducing flexion, compression and extension type injuries (Left: flexion position; Middle: stiff position; Right: extension position).

To mimic the test condition, the FE neck model was fixed at the T1 (the first thoracic vertebra) inferior surface using the casting model as shown in Figure 2a-2b. The integrated neck-head model developed in neural position was rearranged to the stiff axis (Figure 2a) by levelling 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 (Figure 1b), 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.

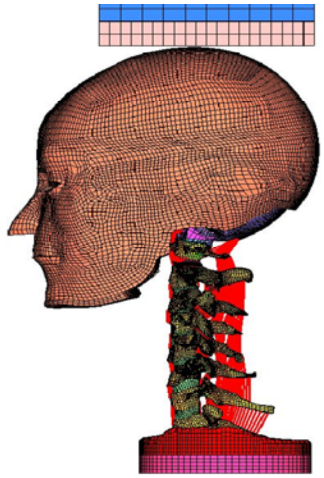

Figure 2a : FE model setup.

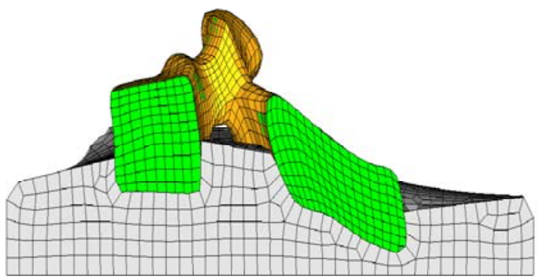

Figure 2b : T1 fixed to the casting in the simulations.

Head-neck complex model

The actual geometry[7] of an average adult male subject (35 years, 1.70 m, and 65 kg) was obtained to construct a high resolution neck FE model with 8-node hexahedral elements representing solid mesh to assure the accuracy and convergence of the tissue (i.e. facet capsule, ligaments) strain/stress calculation in large deformation. Only the passive effects of the head and neck muscles were considered. The Wayne State University Head Injury Model (WSUHIM), previously developed by Zhang et al (2001)[8], was incorporated into the neck model.

Data from head drop tests performed at Duke University[9] and data from rear-end impact sled tests at Wayne State University[10] were used to validate the head-neck complex FE model (Figure 3). The development and validation of this head-neck complex FE model will be presented in another paper.

Relative facet displacement calculation technique

Having validated force-deflection responses of the FE neck model against results from cervical axial compression tests performed by Pintar et al.[6], the compression and facet strain from all three loading conditions (flexion, stiff 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 (Figure 4). 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).

RESULT AND DISCUSSION

Compressive neck injury simulations

In[6] twenty human cadaveric head-neck complexes were tested using a crown impact to the head at speeds from 2.5 m/s to 8.0 m/s. Those cervical axial compression experiments done by The Medical College of Wisconsin are very different from the drop test[9]. 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 (Figure 1a, 2a). 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 head-neck complex model was rearranged to the stiff axis by bringing the vertebrae to the stiff axis by levelling 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.

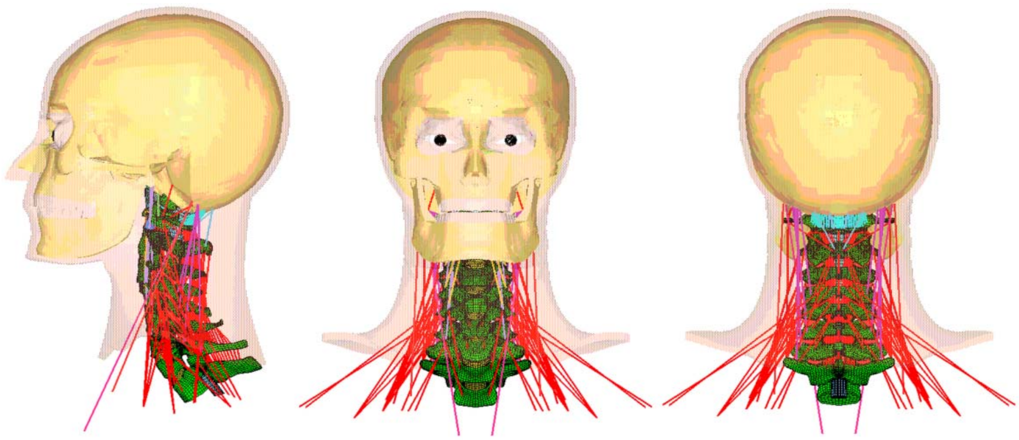

Figure 3 : Integrated 50th percentile human head-neck model with various ligaments and joint capsule and muscles.

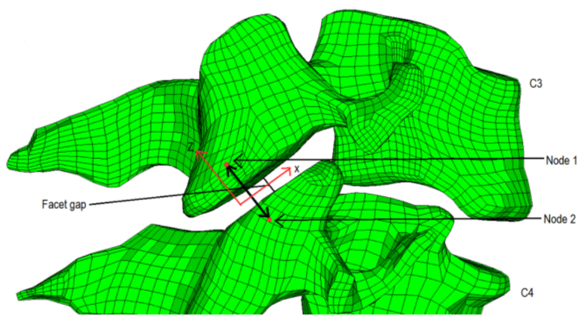

Figure 4: Local coordinate system of facet joint.

A constant velocity of 5.25 m/s was used because the Pintar's paper[6] 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 Pintar's paper.

Besides, 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.

Force responses

The FE model force response for 3.25 m/s and 5.25 m/s was in the range of force versus deformation plots from the Pintar's experiments (Figure 5). 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, the authors 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.

Neck injury patterns

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 Pintar's paper[6].

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.[6] 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.

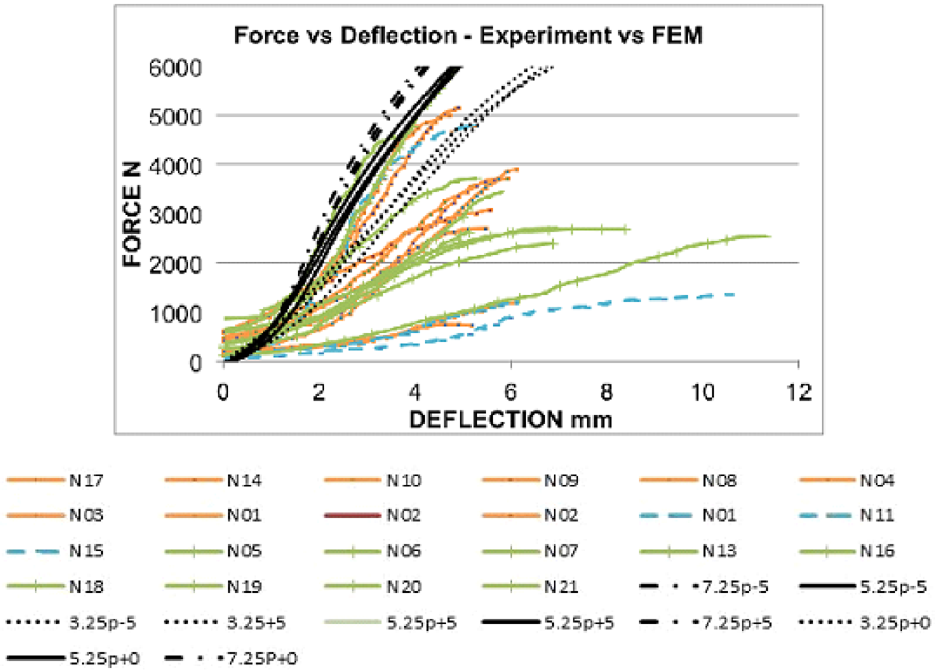

Figure 5 : Model predicted neck force-deflection vs experiment results in[6].

Facet joint injury simulations

At 5.25 m/s impact for example, in case when 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, while the C7-T1 facets sustained tension along X axis. The facets sustained compression in both X and Z axes, and 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 facets which 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 (Figure 6a-6b). 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. 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.

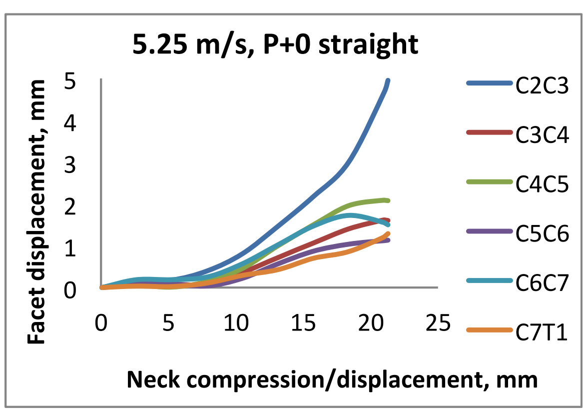

Figure 6a : Facet resultant displacement as the function of neck compression with more sliding occurs in C2-C3 facet in straight neck position from 5.25 m/s impact.

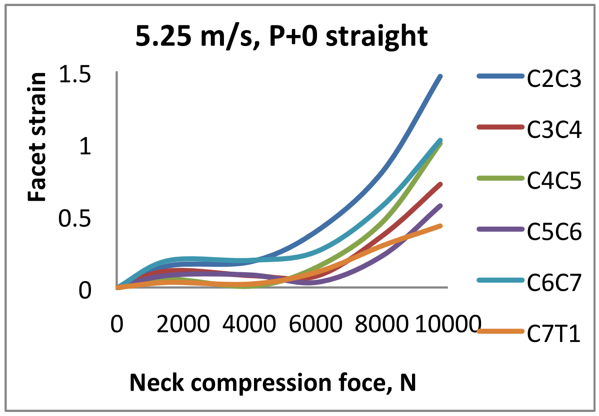

Figure 6b : Facet resultant strain as the function of neck compression force with more sliding occurs in C2-C3 facet in straight neck position from 5.25 m/s impact.

CONCLUSION

Because the FE model force response for 3.25 m/s and 5.25 m/s was in the range of force versus deformation plots from the crown-impact experiments done by Pintar et al, the head-neck complex FE model was further validated.

According to this head-neck complex FE model, compressive neck injuries produced by the crown impact to the head was numerically simulated. There were some interesting findings. For example, 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, and at higher crown-impact velocities the C6-C7 or C7-T1 region was found to be a potential severe injury site. Of course, more simulations of compressive neck injuries from cadaver tests are needed to confirm the suggestions.

ACKNOWLEDGEMENT

This work was supported in part by grants from the National Natural Science Foundation of China (No. 30928005) and the Military Medical Research Foundation of China (Nos. AWS11J008 and BWS12J033).

REFERENCES

[1] A.I.King; Annu.Rev.Biomed.Eng., **2000**, 55-81 **(2000)**.

[2] H.B.Chen, L.Y.Zhang, Z.G.Wang, K.H.Yang, A.I.King; Biomechanics of the Neck, in V.Klika, (Ed); Theoretical Biomechanics., InTech Publisher, Rijeka (Croatia), 385-402 **(2011)**.

[3] N.Yoganandan, S.Kumaresan, F.A.Pintar; Clinical Biomechanics, **16(1)**, 1-27 **(2001)**.

[4] G.S.Bahling, R.T.Bundorf, G.S.Kaspzyk; Rollover and drop tests - The influence of roof strength on injury mechanics using belted dummies, Proceedings of the 34th Stapp Car Crash Conference, Orlando, FL, USA, 4-7 November **(1990)**.

[5] J.W.Hu, K.H.Yang, C.C.Chou, A.I.King; Spine, **33(23)**, 2529-2535 **(2008)**.

[6] F.A.Pintar, N.Yoganandan, L.Voo, J.F.Cusick, D.J.Maliman, A.Sances Jr.; Dynamic characteristics of the human cervical spine, Proceedings of the 39th Stapp Car Crash Conference, San Diego, CA, USA, 8-10 November **(1995)**.

[7] H.B.Chen, L.Y.Zhang, L.W.Tan, S.X.Zhang, X.M.Cheng, Y.Wang, L.Zhang, Z.G.Wang; TELKOMNIKA Journal of Electrical Engineering, **10(8)**, 2320-2329 **(2012)**.

[8] L.Y.Zhang, K.H.Yang, R.Dwarampudi, K.Omori, T.Li, K.Chang, W.N.Hardy, T.B.Khalil, A.I.King; Stapp Car Crash J., **45**, 369-394 **(2001)**.

[9] R.W.Nighitingale, J.H.McElhaney, D.L.Camacho, M.Kelinberger, B.A.Winkelstein, B.S.Myers; The dynamic responses of the cervical spine: Buckling, End conditions and tolerance in compressive impacts, Proceedings of the 41st Stapp Car Crash Conference, Lake Buena Vista, FL, USA, 13-14 November **(1997)**.

[10] B.Deng, P.C.Begeman, K.H.Yang, S.Tashman, A.I.King; Stapp Car Crash Journal, **44**, 171-188 **(2000)**.