The fracture of the human cervical spine

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Abstract

Frequently, such as during automobile collisions, the injury mechanism of the cervical spine is not precisely understood therefore we are not able to establish efficient and robust prevention, diagnostic or treatment methods. Hence, a detailed analysis is still necessary by modelling the spine and its components as accurately as possible with the help of present technical capabilities. In this paper, a preliminary analysis using a 3D finite element model of the cervical spine and the head is presented. In addition, future plans are also presented regarding the topic.

Keywords: cervical spine, finite element analysis, dynamic loads

Introduction

Motivation

In our age, after the automotive industry has largely accelerated transportation in general, a fair number of car accidents occur, some of which result in neck injury. Cervical spine injuries that cause quadriplegia, although not the most frequently occurring injury type, are devastating for the individual as well as for society and, additionally, rather costly. Beside the medical cost related to quadriplegia, there is also significant loss in productivity, both of which is estimated to be $97 billion in the USA,[1] since mostly the young members of the society suffer severe injury. Thus, further investigation is still needed in order to prevent and treat these injuries efficiently.

Injury mechanism types

Injury mechanism categorization may be achieved by several other ways: one of the most popular is based on the global movement of the head relative to the torso that is: compression, tension (or distraction), flexion, extension, rotation and coupled movement of the aforementioned ones.[2,3] However, only a few modes of mechanisms are relevant: compression, compression and flexion, compression and extension, and rotation.[3] It is also worth noting that this classification can be misleading with regards to recognizing the actual injury mechanism. Frequently, the motion of the head is different from the motion of the injured cervical segment. For instance, a flexion motion of the head may be simultaneously present with an extension motion of a spine segment. In addition to that, a local injury of a spine segment may occur before any global head motion is observable.[4]

Compression can lead to a special kind of injury, which is called Jefferson fracture.[5] This injury mechanism is recognizable by the fracture of the anterior and/or posterior arches of the atlas. Another commonly occurring type is burst fracture, which involves the disintegration of one of the vertebral bodies and piercing of the spinal cord by bony fragments.[3] Compression-flexion occurs when an eccentric compressive force acts upon the head, leading to wedge fracture, burst fracture, or anterior dislocation of the cervical vertebrae. In severe cases, dislocation frequently leads to quadriplegia due to greatly injuring the spinal cord. A typical instance of this injury mechanism is the case when the rider is thrown over the vehicle during a motorcycle crash and the head impacts on road surface.[3]

Compression-extension cause injuries to the spinous processes. However, nowadays this type of injury mechanism occurs only when the occupant doesn't use the seat belt. In front-end crashes, the unrestrained occupant slides forward and upward, which can cause the head to extend and impact on the windshield.[3] Tension-extension loading is also a common one, resulting in the Hangman's fracture and disruption of anterior ligaments of the cervical spine. Tension-extension injury mechanism is suffered by, for instance, unbelted occupants whose heads hit the windshield while their torso move forward.[6] A summary and other injury mechanism types are included in the work of Cusick.[2]

Experiments

Experimental investigations are essential in exploring the behavior of the cervical spine under various conditions since these investigations provide validation data for numerical models. Validation of computational models is most commonly based on relatively easily measurable quantities of experiments, such as quasi-static or dynamic global head movement, range of motion of spinal segments due to applied, measured, loads.

The conducted experimental research data are numerous; however, there are only a few type of tests that are most commonly carried out. For instance, one can distinguish between static and dynamic tests. Another categorization might be based on the fact that whether the investigated specimen is alive or not: they are called *in vivo* and *in vitro* tests, respectively. Also, in case of *in vitro* measurements, a further categorization can be made: whole cadaver or segment tests can be conducted. In addition, with regards of the applied load, flexion, extension, lateral bending and axial tests can be distinguished. Beside these types of experiments, there are also range of motion tests and tolerance tests. A few illustrative example follow. A study was conducted to measure cadaver cervical spine tensile tolerance properties.[7] The effect of boundary conditions was also investigated. The rotation-bending moment relationship of the cervical spine is commonly determined.[8] Some researchers investigated even the effects of aging thus degeneration of the spine.[9] Another fairly typical dynamic experimental setting includes a sled upon which a chair is fixed. The sliding board is started at the top of the sled device, which is stopped by a pneumatic cylinder at the bottom. When the deceleration is produced by the pneumatic cylinder, the subject is under a similar condition that is present at vehicular collisions thus the response of the neck can be investigated.[10,11]

FEM neck models

Dynamic neck models often incorporate the whole cervical spine and the head but the accuracy of constitutive and geometrical models are limited. Frequently, vertebral bodies are modelled as rigid bodies and soft tissues as linear springs. However, muscles are typically included in the model. Most papers mention details of the modelling difficulties; here follows by no means an exhaustive list of these hardships. When validating the cervical spine model, a common approach is that the material model characteristics are calibrated so that the numerical model mimic some experimental response. Even though global kinematics can reliably be reconstructed, the problem with calibrating is that the main point of it is to compensate for some modelling deficiency. Thus tissue-level response is likely to be far from biofidelic. To overcome this discrepancy, model development ought to take place at tissue level as far as the geometry and material properties are concerned.[12] In order to account for realistic change in direction of line of action of muscles, intermediate points ought to be inserted, which then constrained to the vertebra, over which it spans.[13,14] This consideration is emphasized by many.[15,16]

Methods

Geometry

In order to produce a biofidelic response, sufficient amount of detail ought to be included in the geometrical model. Fortunately, there has been a huge effort to build a full human body geometry model. The improved work of Mitsuhashi[17] was used to build the geometrical model of the bones.[18]

The general overview of the definition of the geometrical model is as follows. The relevant parts of the skeleton were loaded in Spaceclaim[19] in order to additionally define ligaments and muscles as line bodies in between bones. Then, all the geometrical model of the whole head-neck complex were loaded into Ansys Mechanical.[20]

More precisely, the following bony parts were modelled as solid bodies: **skull without mandible, C1, C2** and **C3 vertebra**. As far as the soft tissues are concerned, the **IVD between the C2 and C3 vertebra**, and **ALL, PLL, LF, ISL, CL, AAAL, PAAL, TL, AAOM, PAOM** and **TM** ligamentous structures, and **MIS, MIT, MR, MOCS** and **MRCPMi** of the muscles were included in the form of line bodies. *(Figure 1)* The cross-sectional areas of these soft tissues were obtained from previous measurements.[21–23] Besides, fictional cartilage was also built in order to establish a simple bonded connection between the skull and the atlas. Another fact worth paying attention to is that the mandible was neglected in order to simplify the meshing process and also reduce the number of finite elements.

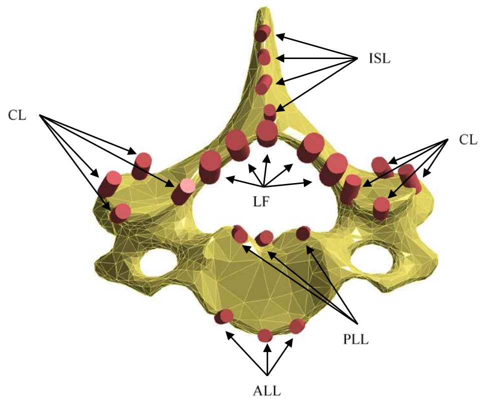


Figure 1. Superior view of C3 with ligaments connecting C3 to C2

Material models

In the finite element model, only homogenous isotropic, linearly elastic material models are applied. The case of soft tissues is an exception: their material law is nonlinear, namely: in tension they follow the linearly elastic material law but in compression they do not exert any force. In case of the modelled bones, calibration of the mass density was necessary. Since the mandible and most of the soft tissues surrounding the skull was ignored, the inertial effect of the whole head was taken into account in an approximate way by changing the mass density of the skull model so that its mass is equivalent to whole head's.[24]

In addition, the overall mass density of the vertebrae was calculated based on the mass density of the two constituent bone tissue.[24,25] The summary of the applied material models are found in *Table 1.*

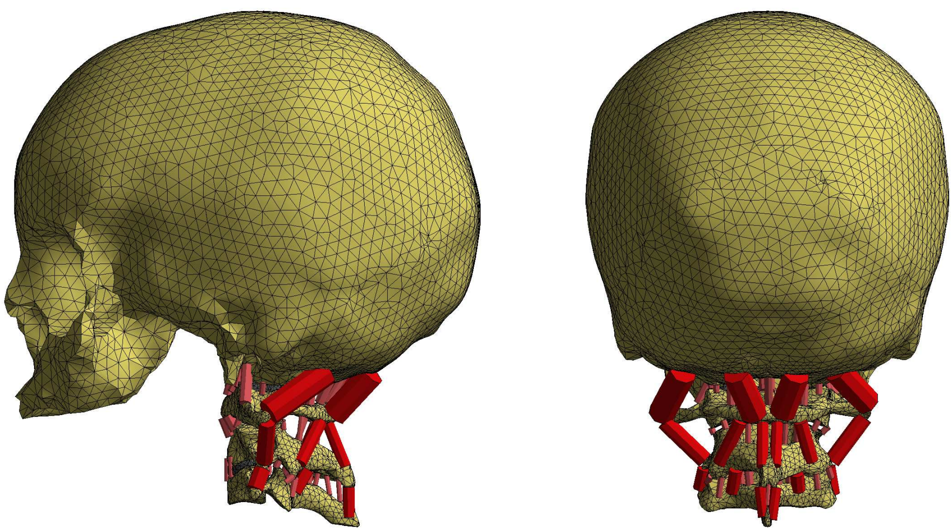
Tissue/Anatomical part	Mass density [g/cm3]	Young's modulus [MPa]	Poisson's ratio [-]
Vertebrae	1,381	18000[26]	0,4[27]
Skull	9,509	18000[26]	0,4[27]
Ligaments	1,1	100[27]	0,4[27]
Intervertebral disc	1,1	100[28]	0,3[28]
Articular cartilages	1,1[26]	10[26]	0,4[27]
Muscles	1,0576[29]	100	0,4[27]

Table 1. Applied material properties of the FE model

Finite element model

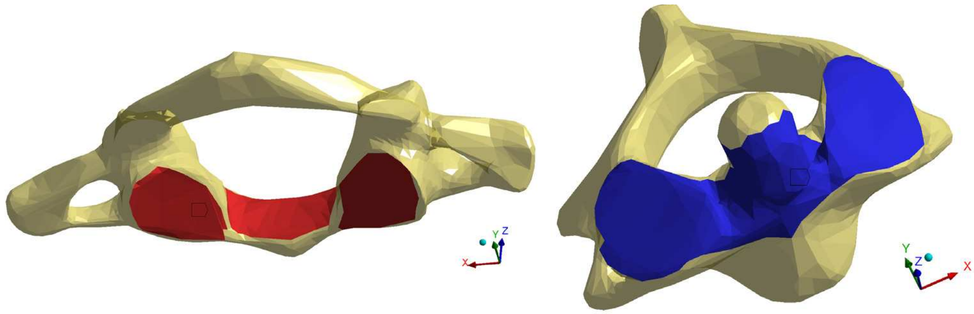
Finite element mesh consists of quadratic tetrahedron elements (element type: SOLID187) and cable elements (element type: LINK180) resisting only axial tensile forces. *(Figure 2)* Element size is influenced by the volume and complexity of the body that is to be meshed. For the skull, the vertebrae and intervertebral disc, the minimum element size is set to 3 mm. In articular cartilage between the occiput and the atlas, the defined minimum element size is 1,5 mm. However, a different consideration was applied to the line bodies in the model: each line body was meshed with only one LINK180 element.

The geometric model, which provided the base for the mesh, is composed of several individual parts, which have no connections geometrical



a) b)

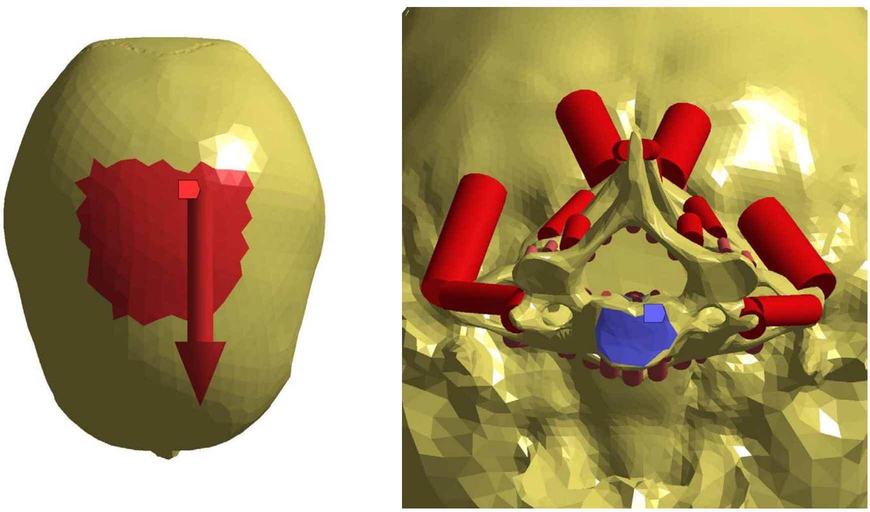
Figure 2. a) Lateral view of finite element model b) Posterior view of the finite element model



a) b)

Figure 3. a) C1 vertebra and its contact surfaces b) C2 and its contact surfaces

model-wise. This means that a key question of developing the finite element model is to establish proper connections between these separate parts. Contact elements were used to establish proper connections between meshed solid bodies. For one of the analyzes that are presented in this paper, all connections are bonded (D1B analysis). In case of the other one, frictionless contact behavior was set between C1 and C2 vertebrae and bonded contact for every other solid body connection (D1F analysis). The contact surfaces of C1 and C2 are shown on *Figure 3*. In case of joining line elements to 3D elements, line element nodes were connected to the nodes of tetrahedrons by several, automatically created beam elements. This connection lets the LINK180 elements to rotate but distributes the axial forces that are transmitted from these LINK180 elements.



a) b)

Figure 4. a) Superior view of the cranium: surface over which the distributed loads are applied (red) b) Inferior view of the model: surface over which the fixed support is set (blue)

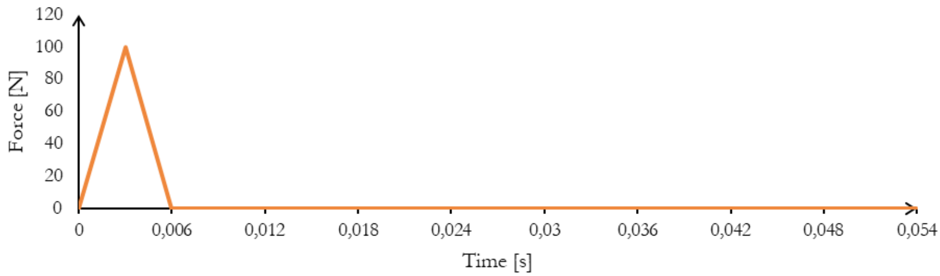


Figure 5. Dynamic force magnitude vs. time diagram

Applied loads and boundary conditions; analyzes

The two analyzes consist of dynamic surface loads and no gravitational load. These surface loads are distributed over the superior part of the skull and have 100 N peak magnitude. In conjunction with these loads, there's a fixed support distributed on the inferior surface of C3. *(Figure 4)*

Results

D1B analysis results

Since the magnitude of the resulting displacement field of the model is quite small, presenting the motion of the head by diagrams showing the displacement component vs. time is much more beneficial. The coordinate system's X, Y and Z axes correspond to the frontal, sagittal and longitudinal axes, respectively. Positive directions are defined in the sinister, posterior and superior directions along the three anatomical axes. Now we can notice that, indeed, the model exhibits a slight asymmetric motion since the X component of the resulting motions are not zero. *(Figure 6)* Besides, the graph suggests that hardly any flexion motion was produced due to the applied distributed force since all component of the displacement of the skull's center of gravity takes on negative values to a very low extent.

Regarding soft tissues, their cable-like behavior can be clearly seen. *(Figure 7 and Figure 8)* In flexion, the posterior ligaments and muscles are in tension while in extension, the anterior soft tissues exert tensile forces. The neck is in flexion or in extension when Y component of displacement of the skull's cen-

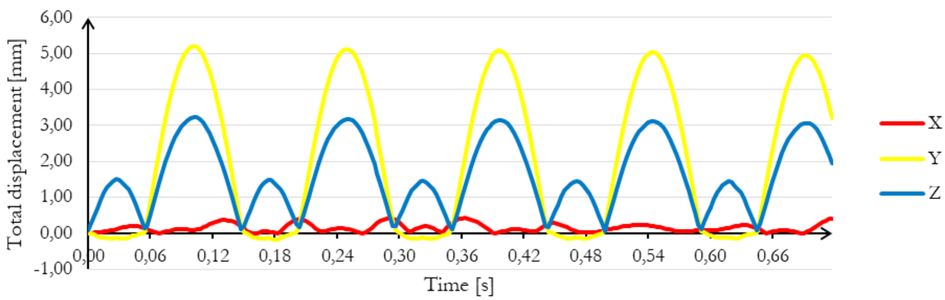


Figure 6. Directional displacement of the skull's center of gravity vs. time during D1B analysis

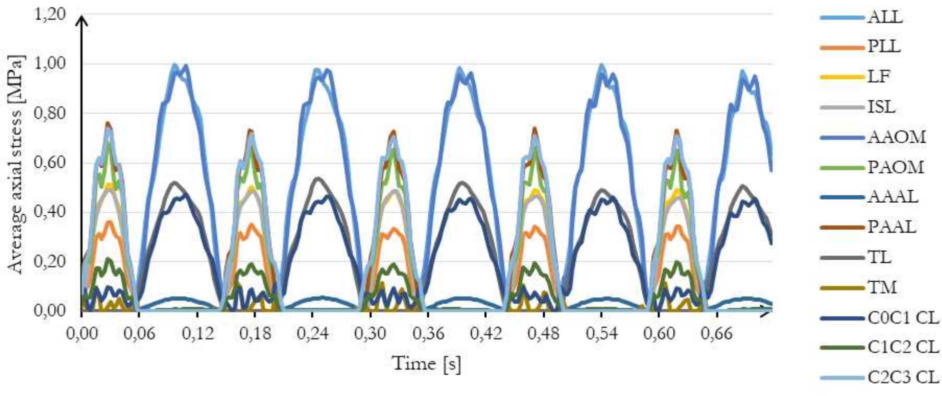


Figure 7. Average axial stress of different ligaments during D1B analysis

ter of gravity takes negative or positive values, respectively.

Turning back to the results of ligaments, one can notice that, apart from the high degree of regularity, in flexion (0,00-0,06 s) more ligaments become tensed than in extension (0,06-0,15 s). This result reflects the fact that, in flexion, the posterior elements of the vertebrae move away from each other and more ligaments connect these posterior elements of adjacent two vertebrae than the vertebral bodies.

As for the muscles, axial tensile stresses are exerted only in flexion and hardly any in extension. Comparing to the case of ligaments, a similar fact may cause this result: only posterior muscles were defined. Considering the results of the muscles, *(Figure 8)* maximum stresses arise in MIS and MR muscles. MIS connects the spinous processes of adjacent vertebrae and MR connects the articular process of one vertebra to the lamina of the superior vertebra. Since the aforementioned bony parts belong to the posterior elements of a vertebra, all of these suggests that relative rotation of two adjacent vertebrae has a greater impact on muscle stresses than the rotation of the skull relative to the spinal column.

D1F analysis results

On *Figure 9*, displacement components may suggest a more realistic motion of the head than in the case of D1B analysis. Now

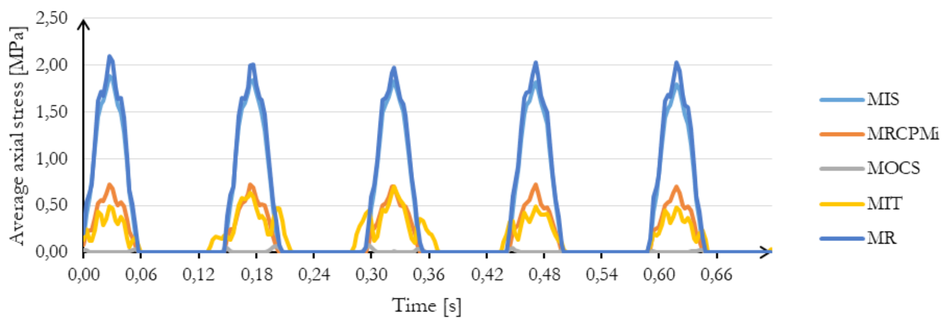


Figure 8. Average axial stress of different muscles during D1B analysis

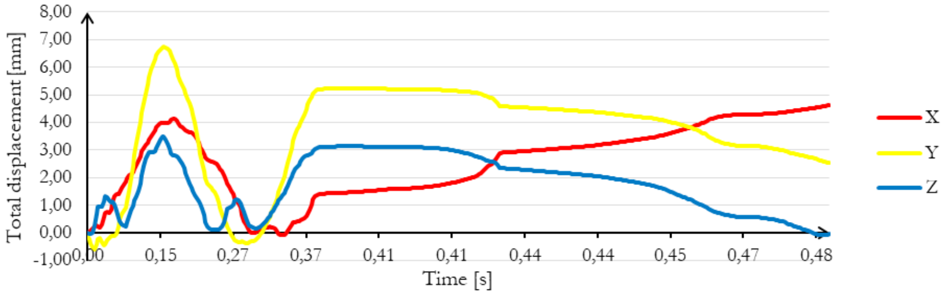


Figure 9. Directional displacement of the skull's center of gravity vs. time diagram during D1F analysis

during the second extension motion of the neck (at around 0,37 s) the cyclicality ceases. Afterwards, a slow sliding into flexion can be observed. This reflects reality in a sense that the cyclicality of the motion of the head due to an impact load does not continue for a prolonged time period.

Where ligaments are concerned, maximum stresses arise, again, in TL. *(Figure 10)* Besides, the same mechanical behavior can be recognized in the soft tissues as in the displacement components: after the second extension motion *(Figure 10, Figure 11)* the cyclicality ends and a slow variation of the stresses begins, which may likely to be in correlation with the motions. As for the muscles, the time instant of the largest flexion motion is clearly indicated by the peak stresses of MIS muscle. The reason is similar as it was in the case of D1B analysis.

Discussion

As a first step of further investigations, a **simplified model of the head-neck complex** was developed that consists of the skull without the mandible, the top three vertebrae, the intervertebral disc between C2 and C3, most of the ligaments, and a few pair of deep muscle.

As far as the top two vertebrae are concerned, a bonded contact is apparently not sufficient of modelling the connection of these vertebrae. Additionally, the articular surfaces of adjacent vertebrae may also likely come in contact with

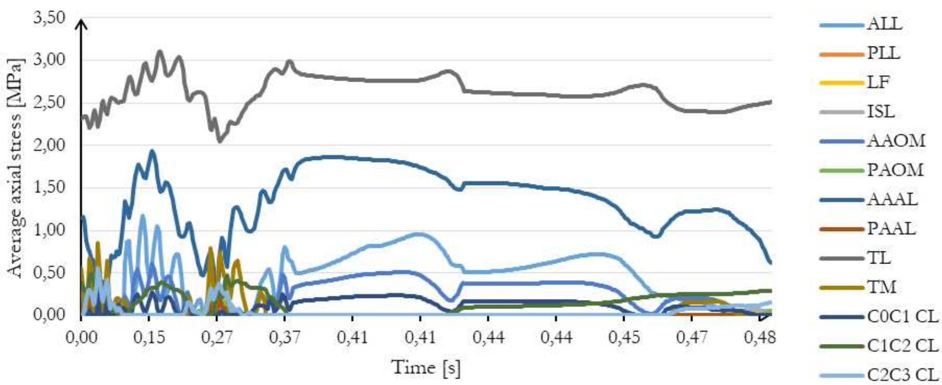


Figure 10. Average axial stress of different ligaments

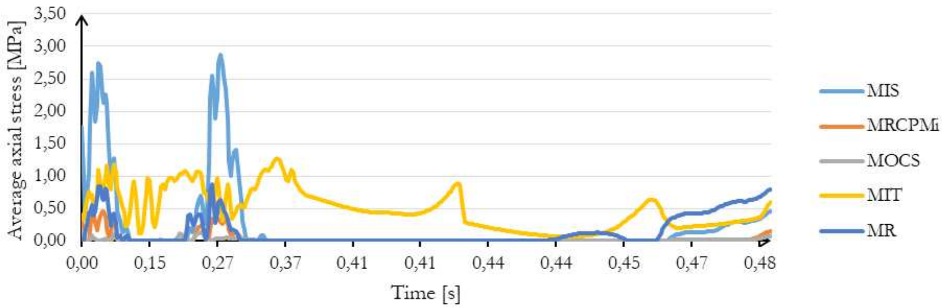


Figure 11. Average axial stress of different muscles

one another therefore at least **frictionless contact** ought to be defined.

One obvious direction of further developing the finite element model is the **incorporation of more vertebrae and the relevant soft tissues.** At least all seven of the cervical vertebrae should be added to the model. The question of the soft tissues are a bit more complex considering the fact that numerous muscles of the back and shoulders have attachment points on one or more of the cervical vertebrae. Proper inclusion of these muscles require special care due to their shape: several of these soft tissues are surface-like thus cannot be modelled as simple 1D line bodies.

Taking into account some of the soft tissues with the help of cable elements may be sufficient in some cases. However, **modelling soft tissues with the help of 2D or 3D elements** would enhance the model's capability of analyzing the response of the neck in much more detail.

Additionally, ligaments and muscles are always in at least modest tension therefore a **pre-stressed state of the soft tissues** would presumably improve the model response. A related question is the determination of proper muscle activation levels that accurately simulate relaxed or tensed neck.

Another important domain of improvement would be the inclusion of **material nonlinearity of soft tissues**. Viscoelastic effects are not negligible therefore accounting for viscoelasticity would affect model response accuracy to a large extent. Taking material nonlinearity into account would be advantageous, especially in case of simulating high speed accidents since viscous effects become more dominant as the loading rate and, consequently, the strain rate increases.

References

1. *French DD, Campbell RR, Sabharwal S, Nelson AL, Palacios PA, Gavin-Dreschnack D.* Health Care Costs for Patients With Chronic Spinal Cord Injury in the Veterans Health Administration. J Spinal Cord Med. 2007;30(5):477–81.
2. *Cusick J, Yoganandan N.* Biomechanics of the cervical spine 4: Major injuries. Clin Biomech. 2002;17(1):1–20.
3. *King AI.* Impact Biomechanics of Neck Injury. In: King AI, editor. The Biomechanics of Impact Injury: Biomechanical Response, Mechanisms of Injury, Human Tolerance and Simulation. Cham: Springer International Publishing; 2018. p. 201–41.

4. *Nightingale RW, Myers BS, Yoganandan N.* Neck Injury Biomechanics. In: The Medical College of Wisconsin Inc on behalf of Narayan Yoganandan, Yoganandan N, Nahum AM, Melvin JW, editors. Accidental Injury. New York, NY: Springer New York; 2015. p. 259–308.
5. *Jefferson G.* Fracture of the atlas vertebra. Report of four cases, and a review of those previously recorded. BR J SURG. 1919;7(27):407–22.
6. *Chen H, Zhang L, Wang Z, Yang K, King A.* Biomechanics of the Neck. In: Klika V, editor. Theoretical Biomechanics. 2011. p. 385–402.
7. *Dibb A, Nightingale R, Luck J, Carol Chancey V, E Fronheiser L, S Myers B.* Tension and Combined Tension-Extension Structural Response and Tolerance Properties of the Human Male Ligamentous Cervical Spine. J Biomech Eng.
8. *Goel VK, Clark CR, Gallaes K, Liu YK.* Moment-rotation relationships of the ligamentous occipito-atlanto-axial complex. J Biomech. 1988;21(8):673–80.
9. *Wheeldon JA, Pintar FA, Knowles S, Yoganandan N.* Experimental flexion/extension data corridors for validation of finite element models of the young, normal cervical spine. J Biomech. 2006;39(2):375–80.
10. *Kumar S, Ferrari R, Narayan Y.* Analysis of right anterolateral impacts: the effect of head rotation on the cervical muscle whiplash response. J Neuroengineering Rehabil. 2005;2:11.
11. *Kumar S, Ferrari R, Narayan Y, Vieira E.* Analysis of right anterolateral impacts: the effect of trunk flexion on the cervical muscle whiplash response. J Neuroengineering Rehabil. 2006 May 16;3:10.
12. *Panzer MB, Fice JB, Cronin DS.* Cervical spine response in frontal crash. Med Eng Phys. 2011;33(9):1147–59.
13. *Dibb AT, Narvekar A, Nightingale RW, Myers BS.* A Comparison of Methods for Modeling Neck Muscle Wrapping in Finite Element Models. Proceedings of the Thirty-Fifth Injury Biomechanics Research International Workshop. 2007
14. *Panzer MB.* Numerical Modelling of the Human Cervical Spine in Frontal Impact. University of Waterloo; 2006.
15. *Brolin K, Halldin P, Leijonhufvud I.* The Effect of Muscle Activation on Neck Response. Traffic Injury Prevention. 2005;6(1):67–76.
16. *van der Horst MJ, Thunnissen J, Happee R, Haaster R, Wismans J.* Influence of muscle activity on head-neck response during impact. SAE Technical Paper 973346, 1997
17. *Mitsuhashi N, Fujieda K, Tamura T, Kawamoto S, Takagi T, Okubo K. BodyParts3D:* 3D structure database for anatomical concepts. Nucleic Acids Res. 2009 Jan;37(Database issue):D782–5.
18. *Hamer S.* Human Anatomy V3 - Male (Life Sciences Japan, BodyParts3D Source). GrabCAD. 2018. Available from: https://grabcad.com/library/human-anatomy-v3-male-life-sciences-japan-bodyparts3d-source-1
19. ANSYS SpaceClaim. Canonsburg, Pennsylvania, USA: Ansys, Inc.; 2018.
20. ANSYS Mechanical. Canonsburg, Pennsylvania, USA: Ansys, Inc.; 2018.
21. *Yoganandan N, Kumaresan S, Pintar FA.* Geometric and Mechanical Properties of Human Cervical Spine Ligaments. J Biomech Eng. 2000;122(6):623–9.
22. *Mattucci SFE, Moulton JA, Chandrashekar N, Cronin DS.* Strain rate dependent properties of human craniovertebral ligaments. J Mech Behav Biomed Mater. 2013 Jul 1;23:71–9.
23. *Borst J, Forbes PA, Happee R, Veeger D (H. EJ).* Muscle parameters for musculoskeletal modelling of the human neck. Clin Biomech. 2011 May 1;26(4):343–51.
24. *Clauser C, McConville J, Young JW. Weight,* Volume, and Center of Mass of Segments of the Human Body. Ohio: Air Force Systems Command Wright- Patterson Air Force Base; 1969 p. 112. Report No.: AD-710 622.
25. *Pan C-Y, Liu P-H, Tseng Y-C, Chou S-T, Wu C-Y, Chang H-P.* Effects of cortical bone thickness and trabecular bone density on primary stability of orthodontic mini-implants. J Dent Sci. 2019 Jul 20;
26. *Pal S.* Mechanical Properties of Biological Materials. In: Design of Artificial Human Joints &

Organs. New York: Springer; 2014.
27. *Korhonen RK, Saarakkala S.* Biomechanics and Modeling of Skeletal Soft Tissues. In: Klika V, editor. Theoretical Biomechanics. 2011.
28. *Meyer F, Bourdet N, Deck C, Willinger R, Raul JS. Human Neck Finite Element Model Development and Validation against Original Experimental Data. Stapp Car Crash J. 2004 Nov;48:177-206*
29. *Klein Breteler MD, Spoor CW, Van der Helm FCT. Measuring muscle and joint geometry parameters of a shoulder for modeling purposes. J Biomech. 1999;32(11):1191–7.*

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