Cervical Spine Injuries -

Numerical Analyses and Statistical Survey

Karin Brolin

Doctoral Thesis

Report 2002-29

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Preface

The work presented in this doctoral thesis has been carried out at the Department of Aeronautics at the Royal Institute of Technology (Stockholm, Sweden). Hans von Holst and Peter Halldin have initiated the research. The two epidemiological studies of neck injuries in Sweden were funded by the Department of Aeronautics. The three numerical studies were collaborations between the Department of Aeronautics, the Department of Neurosurgery at the Karolinska Hospital (Stockholm, Sweden), and Volvo Car Corporation (Göteborg, Sweden).

I wish to thank my supervisor professor Hans von Holst and my colleague Peter Halldin for their guiding my research efforts and for their energy and enthusiasm. I would like to recognize professor Jan Bäcklund for sharing his expert knowledge. Also, I would like to thank the rest of "the Neuronics", Svein Kleiven and Magnus Aare, who have offered support and encouragement when I needed it. I have enjoyed working with you! I would also like to acknowledge my colleagues at the Department of Aeronautics for helping me out in thousands of different ways and cheering up the tea breaks. Also, my sincere thanks to everyone that I've been in touch with during my research, among others Camilla Palmertz, Stefan Larsson, and Kristina Lundgren at Volvo Car Corporation, all the nice people at Alfgam Optimering AB, and of course the Master Thesis students Anna, Erik, Martin, Ingrid, and Mathias.

Finally, I appreciate all the support I've got from my sisters, Elin and Lisa. I thank Magnus, all my friends, my parents, and my poodles for forcing me to relax from work and enjoy life!

Karin Brolin

Stockholm, August 2002

Sammanfattning

Nackskador är en viktig grupp skador eftersom de potentiellt kan orsaka ryggmärgskada. Det kan i svåra fall ge upphov till livshotande eller allvarliga men. Snabba fordon och äventyrliga fritidsaktiviteter har ökat medvetenheten om nackskadors konsekvenser. Statistiska studier fyller en viktig uppgift för att definiera skadornas utbredning och problemets storlek. Utveckling av kraftfulla tekniska och kliniska verktyg är viktig för prevention av nackskador.

Därför är syftet med denna teknologie doktorsavhandling:
- att illustrera utbredningen av nackskador i Sverige,
- att utveckla en finitelement (FE) modell av övre nackkotpelaren och
- att studera stabilitet och instabilitet i nackkotpelaren.

Två statistiska studier med data från Socialstyrelsen har genomförts. All data från slutenvården på svenska sjukhus under 13 år, från 1987 till 1999, studerades. Under den här tidsperioden inträffade 14.310 nackskador varav 782 hade dödlig utgång. För hela populationen var den vanligaste skadenivån i nedre nackkotpelaren. Med ökande ålder flyttar skadenivån uppåt så att den övre nackkotpelaren är det vanligaste skadeområdet för svenskar äldre än 65 år. Under perioden har antalet nackfrakturer minskat för alla åldersgrupper utom för personer äldre än 65 år. Män är överrepresenterade bland patienter med nackfrakturer, vilket gäller oberoende av ålder. Antalet skador orsakade av transportolyckor har sjunkit, vilket medför att fallolyckor nu är den största orsaken bakom nackskador.

En anatomiskt detaljerad FE-modell av den mänskliga nacken har utvecklats. Modellen har jämförts med experiment för att verifiera realistiska rörelser i nackens leder och den överensstämmer väl för extension, flexion, lateralböjning, axiell rotation och dragprovning. FE-modellen visade sig väl lämpad för att simulera den komplexa anatomin och interaktionen i övre nackkotpelaren. Tre FE-studier har genomförts med nackmodellen. I den första studien kombinerades nackmodellen med en huvudmodell och användes för att studera ett innovativt innertak i bilar avsett att skydda passagerarna vid voltningsolyckor. I den andra studien användes FE-modellen av övre nackkotpelaren i en parameterstudie där ligamentens materialegenskaper studerades. I övre nackkotpelaren styrs rörligheten enbart av ligament, som både skall bevara nackens stabilitet och möjliggöra för stora rörelser i lederna. Slutligen användes FE-modellen för att studera olika ligamentskador och deras effekt på stabiliteten i flexion, extension och lateralböjning. Nackmodellen modifierades så att avslitna ligament kunde simuleras för alla de olika strukturerna i övre nackkotpelaren. Studien visade att anterior atlantooccipitalmembranet, ligamentum flavum och capsularligamenten hade störst inflytande i flexion medan anterior longitudinalligamentet och apikalligamentet påverkade extension.

Avhandlingens slutsats är att nackskador i Sverige är ett problem som måste angripas med nya preventiva strategier. Det är viktigt att resultat från forskning om hur fallolyckor i den äldre befolkningen kan minskas implementeras i preventiva program. Avhandlingen har även visat att den utvecklade FE-modellen är ett kraftfullt verktyg i utveckling och utvärdering av preventiva system. FE-modeller kommer att vara viktiga när framtida preventiva strategier skall definieras. Slutligen kan FE-modeller öka den biomekaniska förståelsen för komplexa förlopp i nacken och bidraga till ökad kunskap vid analyser av stabilitet i nacken.

Abstract

Injuries to the neck, or cervical region, are very important since there is a potential risk of damage to the spinal cord. Any neck injury can have devastating if not life threatening consequences. High-speed transportation as well as leisure-time adventures have increased the number of serious neck injuries and made us increasingly aware of its consequences. Surveillance systems and epidemiological studies are important prerequisites in defining the scope of the problem. The development of mechanical and clinical tools is important for primary prevention of neck injuries.

Thus, the main objectives of the present doctoral thesis are:

- To illustrate the dimension of cervical injuries in Sweden,
- To develop a Finite Element (FE) model of the upper cervical spine, and
- To study spinal stability for cervical injuries.

The incidence studies were undertaken with data from the injury surveillance program at the Swedish National Board of Health and Welfare. All in-patient data from Swedish hospitals, ranging over thirteen years from 1987 to 1999, were analyzed. During this period 14,310 non-fatal and 782 fatal cervical injuries occurred. The lower cervical spine is the most frequent location for spinal trauma, although, this changes with age so that the upper cervical spine is the most frequent location for the population over 65 years of age. The incidence for cervical fractures for the Swedish population decreased for all age groups, except for those older than 65 years of age. The male population, in all age groups, has a higher incidence for neck fractures than females. Transportation related cervical fractures have dropped since 1991, leaving fall accidents as the sole largest cause of cervical trauma.

An anatomically detailed FE model of the human upper cervical spine was developed. The model was validated to ensure realistic motions of the joints, with significant correlation for flexion, extension, lateral bending, axial rotation, and tension. It was shown that an FE-model could simulate the complex anatomy and mechanism of the upper cervical spine with good correlation to experimental data. Three studies were conducted with the FE model. Firstly, the model of the upper cervical spine was combined with an FE model of the lower cervical spine and a head model. The complete model was used to investigate a new car roof structure. Secondly, the FE model was used for a parameter study of the ligament material characteristics. The kinematics of the upper cervical spine is controlled by the ligamentous structures. The ligaments have to maintain spinal stability while enabling for large rotations of the joints. Thirdly, the FE-model was used to study spinal injuries and their effect on cervical spinal stability in flexion, extension, and lateral bending. To do this, the intact upper cervical spine FE model was modified to implement ruptures of the various spinal ligaments. Transection of the posterior atlantooccipital membrane, the ligametum flavum and the capsular ligament had the most impact on flexion, while the anterior longitudinal ligament and the apical ligament influenced extension.

It is concluded that neck injuries in Sweden is a problem that needs to be address with new preventive strategies. It is especially important that results from the research on fall accidents among the elderly are implemented in preventive programs. Secondly, it is concluded that an FE model of the cervical region is a powerful tool for development and evaluation of preventive systems. Such models will be important in defining preventive strategies for the future. Lastly, it is concluded that the FE model of the cervical spine can increase the biomechanical understanding of the spine and contribute in analyses of spinal stability.

Dissertation

This doctoral thesis includes a short introduction to spinal anatomy, neck injuries, and to finite element models of the cervical spine. As a part of this thesis, a finite element model of the upper cervical spine was developed. Five appended papers follow the introduction.

PAPER A

Karin Brolin, Hans von Holst. *Cervical Injuries in Sweden, a National Survey of Patient Data from 1987 to 1999.* (Published in Injury Control and Safety Promotion, 9(1): 2002).

PAPER B

Karin Brolin. *Neck Injuries Among the Elderly in Sweden.* (Accepted by Injury Control and Safety Promotion, August 20, 2002)

PAPER C

Peter Halldin, Lotta Jakobsson, Karin Brolin, Camilla Palmertz, Svein Kleiven, Hans von Holst. *Investigation of Conditions that Affect Neck Compression-Flexion Injuries Using Numerical Techniques.* (Published in the 44:th STAPP Car Crash Journal, 2000-01-SC10, 2000).

PAPER D

Karin Brolin, Peter Halldin. *Development of a Finite Element Model of the Upper Cervical Spine and a Parameter Study of Ligament Characteristics.* (Submitted to SPINE, 2002)

PAPER E

Karin Brolin. *Finite Element Analyses of Spinal Stability and Instability due to Ligament Ruptures* (Manuscript, selected results were presented at the IV World Congress of Biomechanics in Calgary, August 5-9, 2002)

Division of Work Between Authors

PAPER A

Karin Brolin, Hans von Holst. *Cervical Injury in Sweden, a National Survey of Patient Data from 1987 to 1999.* (Published in Injury Control and Safety Promotion, 9(1): 2002).

K Brolin made the outline, the work and writing of the paper under the supervision of H von Holst.

PAPER C

Peter Halldin, Lotta Jakobsson, Karin Brolin, Camilla Palmertz, Svein Kleiven, Hans von Holst. *Investigation of Conditions that Affect Neck Compression-Flexion Injuries Using Numerical Techniques.* (Published in the 44:th STAPP Car Crash Journal, 2000-01-SC10, 2000).

P Halldin and H von Holst developed the idea of the EFIS roof. K Brolin did all the modeling work of the neck together with P Halldin, while S Kleiven modeled the head. K Brolin performed the local validation of the upper cervical spine model. P Halldin and K Brolin wrote the paper under the supervision of Camilla Palmertz, Lotta Jacobsson, and Hans von Holst.

PAPER D

Karin Brolin, Peter Halldin. *Development of a Finite Element Model of the Upper Cervical Spine and a Parameter Study of Ligament Characteristics.* (Submitted to SPINE, 2002)

K Brolin and P Halldin outlined the work. K Brolin did most of the modeling and performed all the numerical analysis, under the supervision of P Halldin. K Brolin wrote the paper.

Contents

1 Introduction

Neck injuries are very severe types of injuries, since there is a high risk of fatality or paraplegia. Any neck injury can have devastating if not life threatening consequences. High-speed transportation as well as leisure-time adventures have increased the number of serious neck injuries and made us increasingly aware of the consequences. Many protective devises have been developed to shield the human neck when we practice motor vehicle sports, skiing, mountain climbing and in other dangerous situations. Unfortunately there is a lack of biomechanical data and understanding of injury mechanisms, and therefore these safety devises are not optimal.

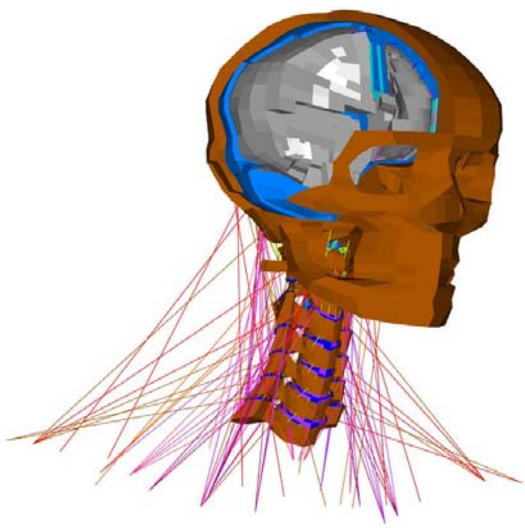
The cervical spine is a complex mechanism from a mechanical and structural point of view. The vertebral column has two major functions, to stabilize the head and to protect the spinal cord. Therefore, cervical spinal injuries are a potential threat to the spinal cord and must be treated with respect and caution. The effects of spinal cord injuries are serious, as mentioned above, ranging from death to quadriplegia, loss of sensory functions, and a range of deficits. A superior location of the spinal cord injury causes more severe effects than an inferior injury. It is therefore of uttermost importance to protect patients with neck injuries from further damage of the spinal cord.

Head and facial trauma is often coupled to serious spinal trauma, Yoganandan et al [1989a]. In order to account for the interaction between the head and the spine, it is a necessary to analyze the head, brain, and cervical spine as one complex. Cervical spinal injuries are caused by impacts going beyond the tolerance threshold of ligaments and bone in the area. The medical personnel would benefit from knowledge of the forces or the kinetic energy causing a fractured cervical spine to find secondary injuries, predict late symptoms and to chose the appropriate treatment. Today, advanced imaging techniques, such as computer tomography (CT) and magnetic resonance tomography (MRT), are good diagnostic tools and can give information on location and type of injury. Medically, it is not always possible to evaluate the stability of the injured cervical spine. Hence, there is a need for better knowledge about how fractures and soft tissue injuries influence the spinal kinematics, as well as injury mechanics.

Researchers today explore the area of spinal injuries with different approaches. Statistical surveys are important tools to define the populations at risk, the external causes of injury, and to evaluate preventive measures. Clinical studies of fractures and follow-up of patients give valuable insight into fracture mechanics and severity of the injury, but lack detailed information of the load situation. Experimentally induced fractures are ideal for studies of injury mechanics and initialization, as the loads applied can be controlled. Unfortunately, there are difficulties in applying realistic load conditions and experimental injuries may not be representative of real life injuries. Numerical analysis is another method with potential to simulate real life accidents better than experimental set-ups. This approach has the advantage of highly detailed models, where stresses and strains can be studied through out the simulation for all the different tissues. One of the drawbacks with numerical analysis is the uncertainty of material properties of biological tissues. All these different approaches are important and when interacting they offer a great opportunity to increase the knowledge of neck injury mechanics.

2 Main Objectives of the Thesis

The main objective of this thesis is to increase the understanding of cervical spinal injuries, treating both injury mechanisms and the influence of injuries on spinal behavior. This is achieved through a combination of epidemiological studies and numerical analyses. A finite element model of the cervical spine was developed to enable detailed studies of cervical kinematics, as well as tissue stresses and strains. Special effort was put on modeling of the upper cervical spine, to ensure a realistic coupling between the cervical spine and the head.



3 Anatomy of the Cervical Spine

The spinal column is a complex structure, whose main functions are to hold us upright, support the head, and protect the spinal cord and blood vessels. The spine is divided into five regions, Figure 1:

- The cervical spine,
- The thoracic spine connecting to the ribs,
- The lumbar spine,
- The sacrum, and
- The coccyx.

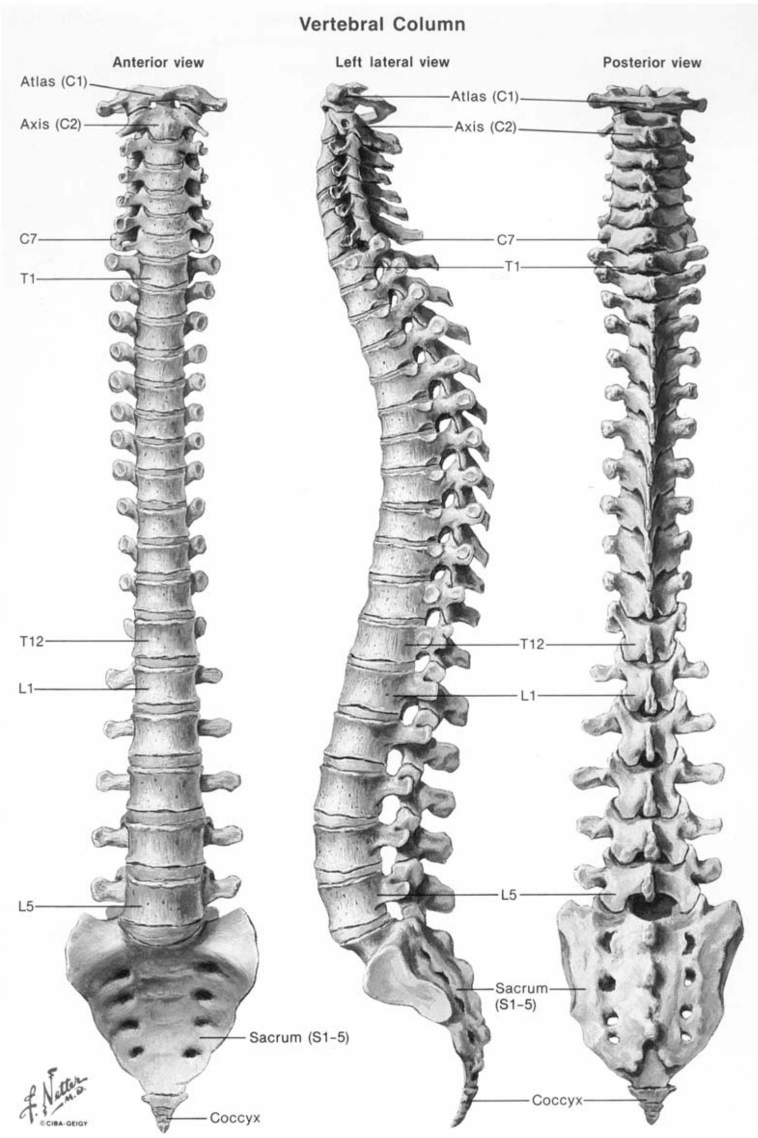


Figure 1: The human spine with the cervical spine extending from C1 to C7, the thoracic spine from T1 to T12, the lumbar spine from L1 to L5, the sacrum and the coccyx. (Copyright 1962 & 1992. Icon Learning Systems, LLC, a subsidiary of MediMedia USA Inc. Reprinted with permission from ICON Learning Systems, LLC, illustrated by Frank H. Netter, MD. All rights reserved.)

In the neck, the spinal region is called the cervical spine. It contains the top seven vertebrae. The cervical spine supports the head and allows a wide range of head motion. It can be divided into the upper and lower cervical spine. This report focuses mainly on the upper cervical spine, which has a different anatomy than the rest of the vertebral column and is the most flexible part of the cervical spine.

There are several different systems of naming the vertebrae. In this report the following is used: the top vertebra is C1, the second C2 and so on until the last cervical vertebra, C7. The occiput, or head, is called C0, even though it is not part of the spine. The first two vertebrae are called the atlas and axis. C1 is named atlas after the giant who carried the earth on his shoulders; similarly the atlas holds up the head. C2 is named axis, probably because it provides the axis for axial rotation in the upper cervical spines. Two consecutive vertebrae, the connecting joints and intermediate ligaments make up a motion segment. Hence, the cervical spine contains eight motion segments. These are named after the two vertebrae: C0/C1, C1/C2, etcetera. The last motion segment is C7/T1, where T1 is the first thoracic vertebra.

Medical terminology used in this chapter is explained in section Terminology, page 28. Special notice should be paid to the word extension, which technically and medically has two different meanings. In this report the medical interpretation is used, that is, the neck is bent backwards (see section on Terminology).

3.1 The Vertebrae

The vertebral bone is of a sandwich structure. It has a stiff outer shell, the cortical bone, and a porous inner marrow, the trabecular bone. The geometry of the vertebrae in the upper cervical spine, C1 and C2, is significantly different from the lower cervical spine, C3 to C7, Figure 2.

3.1.1 The Upper Cervical Spinal Vertebrae

The upper cervical spine is made up of two vertebrae, the atlas (C1) and the axis (C2). The atlas is composed of a bony ring and lacks a vertebral body. It is divided into the anterior and posterior arch. On the lateral masses of the atlas there are facet joint surfaces, Figure 2. The superior facets connect with the occipital condyles (two bony protuberances on the base of the skull) and make up the occipitoatlantal joint. The lower facet joint surfaces match the superior facets of the axis. The axis is composed of a vertebral body and posterior arch, just like the lower cervical vertebrae. In addition, the axis also contains the dens or odontoid process that projects superiorly between the lateral masses of the atlas, Figure 2. Two small joint surfaces on either side of the dens are in contact with the anterior arch of the axis and the transverse ligament, respectively.

3.1.2 The Lower Cervical Spinal Vertebrae

A typical vertebra in the lower cervical spine has an elliptical vertebral body and a bony posterior arch, Figure 2. The posterior arch can be divided into the pedicles, the lamina, the spinous process, the transverse process, and the superior and inferior facet joint surfaces. The spinal cord is protected and surrounded by the posterior elements.

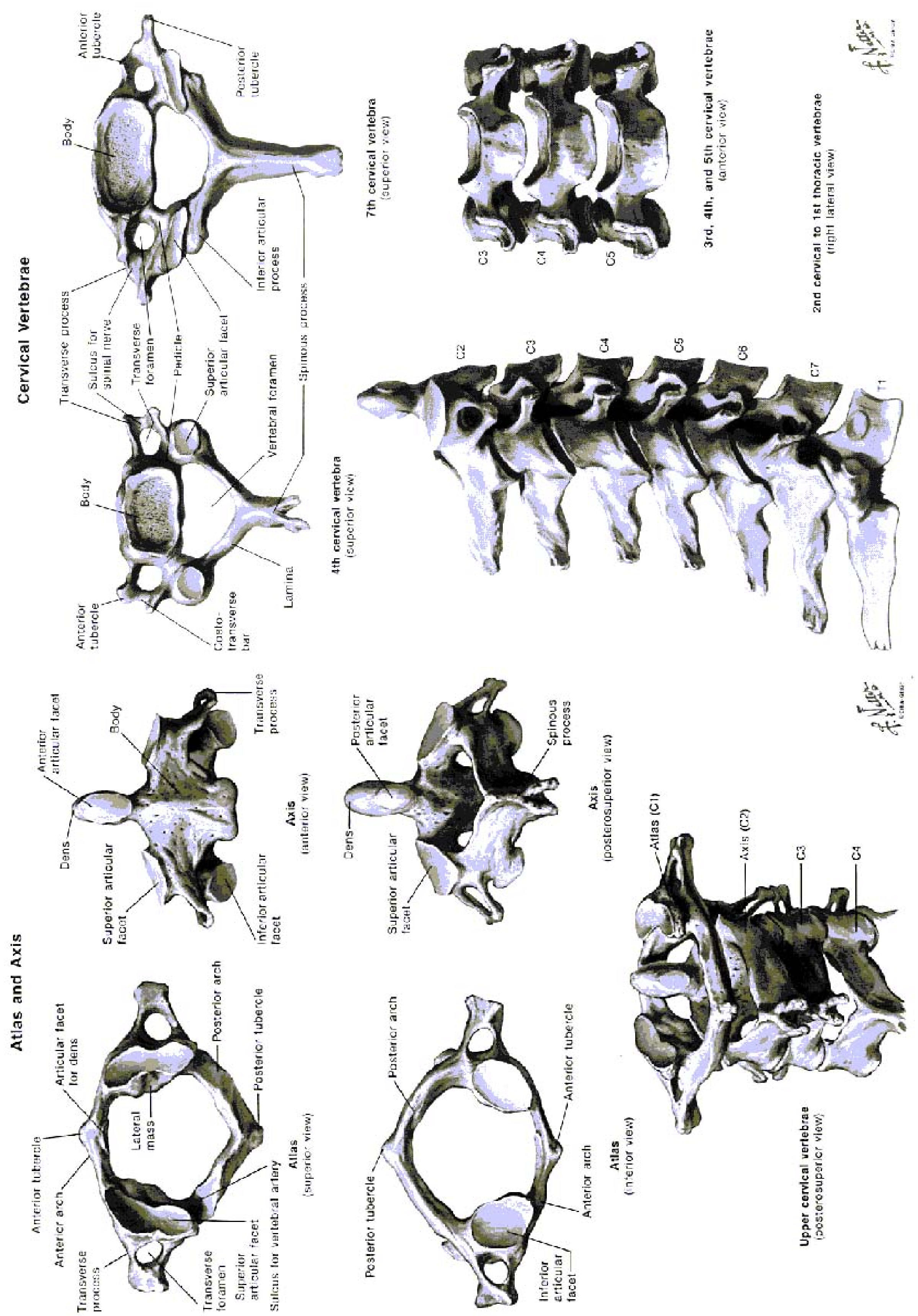


Figure 2: Anatomy of the cervical vertebrae in the upper and lower cervical spine and vertebral column from C1 to C3 and from C2 to T1. (Copyright 1962 & 1992. Icon Learning Systems, LLC, a subsidiary of MediMedia USA Inc. Reprinted with permission from ICON Learning Systems, LLC, illustrated by Frank H. Netter, MD. All rights reserved.)

3.2 *The Joints*

In the upper cervical spine there are two joints, the occipitoatlantal joint and the atlantoaxial joint. These joints differ from the joints in the lower cervical spine since they do not have a disc. The different structures give distinct characteristics. The occipitoatlantal joint is most flexible in flexion-extension, the so-called ‘yes-motion’ of the head. The main function of the atlantoaxial joint is to allow rotation, a ‘no-motion’ of the head. Lateral bending is distributed evenly between the spinal joints. (Myers and Winkelstein [1995]).

3.2.1 The Occipitoatlantal Joint

The occipitoatlantal joint is formed by the occipital condyles on the skull base and the superior facet surfaces of the atlas. The lack of vertebral body on C1 and the shape of the joint surfaces enables considerable mobility in flexion-extension of this joint.

3.2.2 The Atlantoaxial Joint

The atlantoaxial joint is composed of three synovial joints between the atlas and the axis. The superior facet surfaces of the axis and the inferior facet surfaces of the atlas form two facet joints, the third joint is between the anterior arch of the atlas and the dens. This joint enables considerable axial rotation, as the atlas rotates around the dens on the axis.

3.2.3 The Lower Cervical Spine Joints

The joint between C2 and C3 is of the same structure as the joints in the lower cervical spine, that is, two posterior facet joints and an intervertebral disc. The disc is a fibro cartilaginous joint. It has a fluid like central portion (nucleus pulposus) and an outer fibrous, solid structure (annulus fibrosus). The annulus fibrosus is a composite, where the annulus fiber are embedded in a matrix of annulus ground substance.

3.3 *The Ligaments*

Ligaments stabilize the joints in the spine and restrict motion. The lower cervical spinal ligaments are the anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL), the capsular ligaments (CL), the ligamentum flavum (LF), the inter spinous ligaments (ISL), and the super spinous ligaments (SSL) as shown in Figure 3. The upper parts of these ligaments also belong to the upper cervical spine. Ligaments specific to the upper cervical spine are the apical ligament, the alar ligament, the transverse ligament (TL), the tectorial membrane (TM), the anterior atlantooccipital membrane (AAOM), and the posterior atlantooccipital membrane (PAOM), as shown in Figure 4. The TM is the continuation of the PLL from C2 to the occiput, the AAOM the continuation of the ALL from C1 to the occiput. The PAOM is the continuation between C1 and the head for the LF, Myklebust et al [1988]. Several authors have treated the subject of spinal ligaments, among others Myers and Winkelstein [1995], McElhaney and Myers [1993], Panjabi et al [1991c], Saldinger et al [1990], Myklebust et al [1988], and Dvorack et al [1988].

3.3.1 The Apical Ligament

The apical ligament attaches posteriorly at the superior surface of the dens and on the occipital foramen, Figure 4A. It is thin and slightly V-shaped, with the majority of the fibers concentrated in the middle, Panjabi et al [1991c]. The purpose of the apical ligament is to restrain flexion.

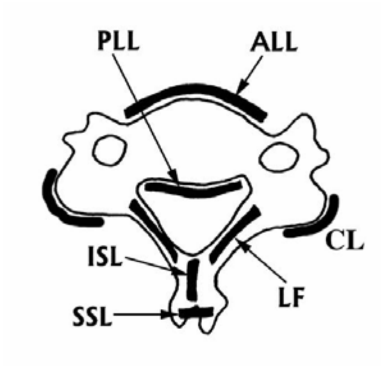
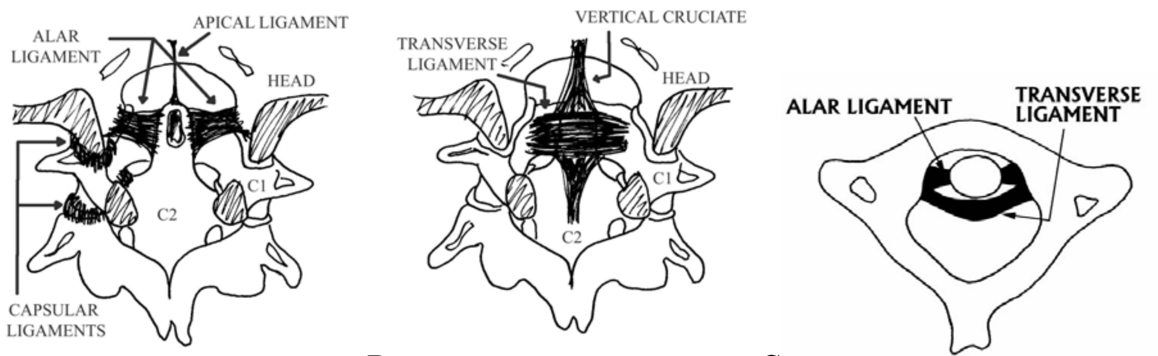


Figure 3: Lower cervical spine ligaments.



A B C

Figure 4: Upper cervical spine ligaments. A) Posterior view of occiput to C2 with apical and alar ligament. B) Posterior view of occiput to C2 with transverse membrane and vertical cruciate. C) Superior view of C1 with alar and transverse ligaments.

3.3.2 The Alar Ligament

The alar ligaments restrain axial rotation of the occipitoatlantal joint, Figure 4A and Figure 4C. Its origin is the lateral margins on the upper third of the dens and the insertion is mainly on the occiput, but also on the lateral masses of C1, Panjabi et al [1991c]. The main fiber constituent is collagen, thus giving an inelastic ligament with a large stiffness, Saldinger et al [1990]. Crisco et al [1991b] developed a model describing how the alar ligaments work. They claimed that both the left and the right alar ligaments limit axial rotation in both directions.

3.3.3 The Transverse Ligament and Vertical Cruciate

The transverse ligament originates on one side of the lateral masses of the atlas and inserts on the other, passing on the posterior side of the dens (middle and upper thirds), Figure 4B. It works as a restraining band on the dens, holding it against the anterior ring of the atlas, thus preventing movement of the dens toward the spinal cord, Panjabi et al [1991c]. The transverse ligament has two vertical extensions: an upward prolongation to the occiput and a downward extension to the vertebral body of the axis. These vertical fibers are called the vertical cruciate and restrain flexion of the head, Dvorak et al [1988]. The transverse ligament fibers are mainly of collagen as the alar ligaments, Saldinger et al [1990].

3.3.4 The Anterior Longitudinal Ligament and the Anterior Atlantooccipital Membrane

The ALL tightly adheres to the anterior surface of the vertebral bodies and the discs between them, Figure 3. It is a broad ligament that narrows at the C1/C2 level. The ALL is replaced by the anterior atlantooccipital membrane at the C0/C1 level (Myklebust et al [1988]). The superficial fiber layers span several motion segments while the deeper fibers stretch between two adjacent vertebrae (Panjabi et al [1991c]).

3.3.5 The Posterior Longitudinal Ligament and the Tectorial Membrane

The PLL runs on the posterior surface of the vertebral bodies and is the anterior portion of the spinal canal, Figure 3. The PLL extends along the spine up to the C2 vertebra. The TM runs between C2 and the occiput, as a continuation of the PLL.

3.3.6 The Ligamentum Flavum and the Posterior Atlantooccipital Membrane

The ligamentum flavum connects adjacent lamina and is located within the spinal canal, on the posterior surface, Figure 3. Unlike most ligaments it contains a high concentration of elastin fibers, making it a tough ligament.

3.3.7 The Supraspinous and Interspinous Ligaments

The SSL and ISL connect the spinous processes of adjacent vertebrae, Figure 3. They restrain extension of the spine. The SSL is absent in the upper cervical spine, while the ISL has reduced strength and appearance at the top level (C1/C2).

3.3.8 The Capsular Ligaments

The CL connects two adjacent facet joints surfaces, Figure 3 and Figure 4A, and one of their main functions is to keep the synovial fluid inside the joint. The CL is often described as thin and loose, especially in the occipitoatlantal (C0-C1) and atlantoaxial (C1-C2) joints, but they function in restraining rotation of the joints. The CL in the atlantoaxial joint were studied by Crisco et al [1991a] and found to restrain axial rotation.

4 Neck Injuries

There is a lack of epidemiological data for neck injuries. Studies of smaller populations have been conducted. Roberge and Samuels [1999], Lowery et al [2001], and Holly et al [2002] collected data from emergency hospitals. Other studies consider only part of the population, Spivak et al [1994], Browne et al [2001], Patel et al [2001], Lomoschitz et al [2002]. There are a few analyses of neck injuries in traffic accidents, among others Thorson [1972], Huelke et al [1981], Krantz [1985], Yoganandan et al [1989a], Bourbeau et al [1993], Miettinen et al [2002]. So far, national data have not been published on neck injuries. The following chapter is a complement to Papers A and B. These two papers describe the incidence and the external causes of those neck injuries that caused hospitalization from 1987 to 1999, in Sweden.

4.1.1 Background

It has been shown that, cervical spine injuries are more often connected with spinal cord injuries than the lower spinal regions, Yoganandan et al [1989a]. There is also a strong association between head and face trauma and neck injuries, Holly et al [2002]. In the cervical spine, the craniocervical junction (Occiput–C2) and the C5-C6 motion segment are the primary locations of cervical injury, according to Yoganandan et al [1989a]. The craniocervical junction is vulnerable due to the flexibility of the upper cervical spine. The most inferior motion segments have a higher incidence because of the stiffness change between the flexible cervical spine and the stiffer thoracic spine. In Papers A and B it was found that the elderly population had a higher incidence for upper cervical injuries rather than lower cervical injuries.

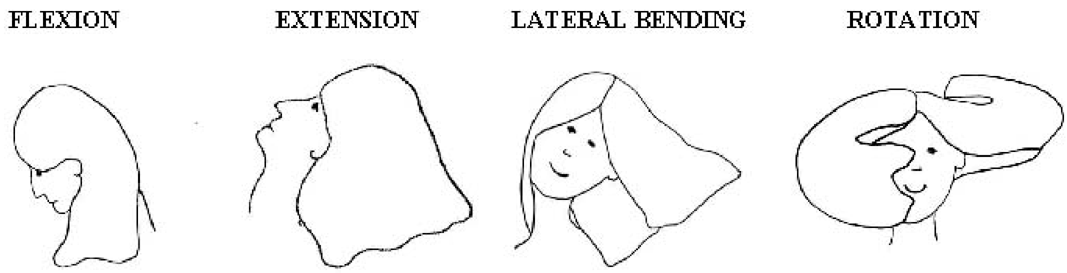


Figure 5: Global motions of the head compared to the torso.

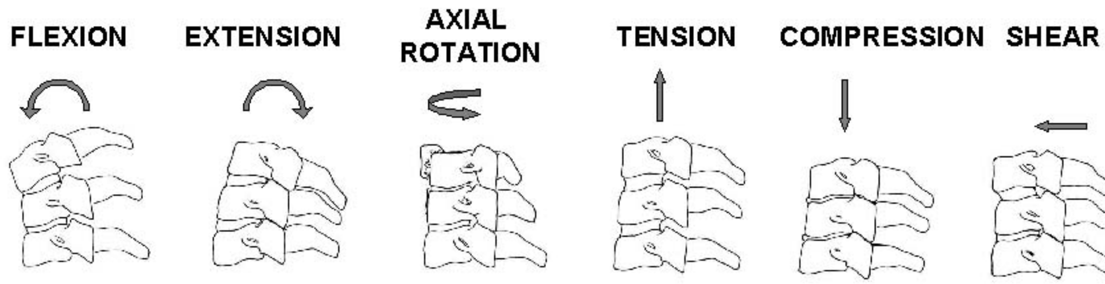


Figure 6: Local motions of lower cervical spine segments, due to different loading modes.

Neck injuries can be divided into two major classes, vertebral fractures and soft tissue injuries (International Classification of Diagnosis version 10). Neck injury classifications usually refers to the global loading mode, which is the motion of the head relative to the torso. This is not always appropriate since the local loading mode between two consecutive vertebrae may differ from the global motion. For example, a compression loading of the head may cause local compression, flexion, and extension loading in different motion segments. Figure 5 defines the terminology for global motions, while Figure 6 illustrates the corresponding local loading modes. Another way of characterizing injuries is by use of the Abbreviated Injury Scaling (AIS), where AIS 0 is non-injury and AIS 6 describes a fatal injury. Clinically, spinal trauma is evaluated in terms of spinal stability. An unstable cervical spine is a threat to the spinal cord and medical treatment is necessary to reduce the risk of spinal cord injury. A stable spine can protect the spinal cord under normal physiologic loads and more conservative treatments can be suitable. It is important to distinguish between the medical interpretation of spinal stability and the mechanical meaning of the word stability. There are two major types of spinal instability, acute and chronic instability. For injuries with acute instability there are immediate risks of spinal cord injury even for small spinal movements. An injury with chronic instability is considered stable from the beginning and develops instability over time, therefore the spinal cord injury may occur many years after the original injury occasion.

This chapter gives a brief overview of some of the more common neck injuries and their injury mechanisms. To correlate with Papers A and B, the injuries are grouped according to their classifications in the Hospital Discharge Register. This presentation does not claim to be complete, since it is not possible to cover all neck injury research in just one chapter.

4.2 Atlas Fractures

4.2.1 Multipart fracture / Jefferson's fracture

A multipart fracture of the first cervical vertebra is often called Jefferson's fracture, although the Jefferson's fracture properly refers to a four-part fracture (McElhaney and Myers [1993]). This injury may be fatal since the nerve roots leaving the spinal canal at this level are in control of the autonomous system, for example heart functions and breathing. Presently, the atlas fractures are classified into three groups. Type I fractures include bilateral single arch fractures, where either the posterior or the anterior arch is intact. Type II fractures, Figure 7, have both posterior and anterior arch fractures, while type III includes fracture of the lateral masses (Lee et al [1998]). Fatalities and instabilities from these types of injuries are common. This fracture type is characterized by displacement of the lateral masses of C1, rather than the number of fractures (Oda et al [1992], Lee et al [1998]). Also, the status of the transverse ligament is important for the stability of these fractures. If the transverse ligament is intact the fracture is clinically considered stable, while a combination of fracture and rupture of the transverse ligament is defined as an instable injury. Lateral mass displacement larger than 5-7 mm is associated with a higher incidence of transverse ligament rupture and spinal instability (Lee et al [1998], Grane [1994], An [1998]). The unstable injury is often treated with immobilization and traction, but there is still a risk for chronic instability (Oda et al [1992]). Atlas fractures are classified as compressive injuries by most authors. An [1998] points out that a severe global extension can compress the posterior arch of the atlas between the skull and the spinous process of C2, thus resulting in a multipart fracture. Other loading modes in combination with compressive loading will result in nonsymmetrical fractures, which is confirmed in a numerical study of C1 fracture patterns by Teo and Ng [2001].

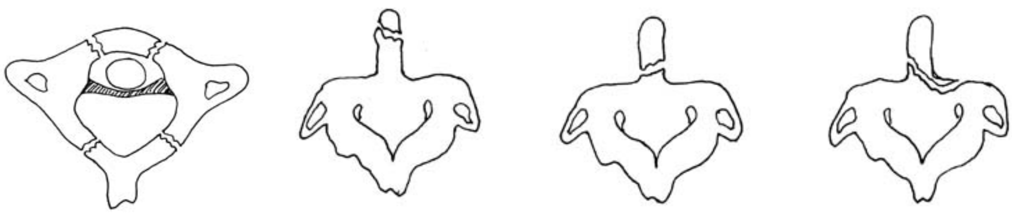


Figure 7: From left to right; Atlas fracture (type II), Dens fractures (type I – III).

4.3 *Axis Fractures*

Axis fractures have a high death rate, where many die at the scene of accident and will therefore be neglected in statistical studies based on hospital reviews (Greene et al [1997]). There are three major types of axis fractures: dens fracture, hangman's fracture, and vertebral body fractures. The dens and hangman's fractures are described below, while the most common vertebral body fractures are described together with the lower cervical spinal injuries in section 4.4.

4.3.1 Odontoid fracture / Dens fracture

The odontoid fracture, or dens fracture, is a fracture of the odontoid process. Odontoid fractures are classified into three major classes: type I, II, and III, illustrated in Figure 7. This fracture type is caused by a local shear loading between C1 and C2. Local shear at this level may rise from global shear or from extreme flexion, extension, or axial rotation. Lateral loading of the dens or the C1, produced either by global lateral bending or axial rotation, consistently results in type II fractures according to Graham et al [2000]. They also found global extension to produce type III fractures. According to Greene et al [1997], the type II dens fracture is the most common fracture of the second cervical vertebra and also the most difficult to treat. Type II fractures are mostly treated with internal fixation. The success of the treatment is related to the relative displacement between the dens and the vertebral body. A factor that decreases the chances of successful treatment is older age (Greene et al [1997]), although this is not agreed upon by all researchers (An [1998]). There are many different techniques for internal fixation, such as wiring, grafts, and screws. This subject is treated by many researchers, among others, Greene et al [1997], Oda et al [1999], Naderi et al [1998], Morandi et al [1999], Fujimura [1996], and An [1998]. Type I fractures are very rare and considered stable (Grane [1994], Greene et al [1997], McElhaney and Myers [1993]). They are usually treated with external fixation. The type III dens fractures are considered more stable than the type II fractures. It is often successful to treat type III dens fractures with external fixations, such as the Halo west (Greene et al [1997], McElhaney and Myers [1993]).

4.3.2 Hangman's fracture

Hangman's fracture, or traumatic spondylolisthesis of the axis, is a fracture of the pedicles or between the facet surfaces (pars interarticularis) in the posterior arch of C2, Figure 8 (Greene et al [1997], McElhaney and Myers [1993]). This injury is thought to be the result of either tension or tension-extension loading, such as blows to the face or, as the name suggests, result of a judicial hanging. It can also be caused by a flexion-compression loading, for example in diving accidents. Most patients surviving this injury are successfully treated with external immobilization (Greene et al [1997]).

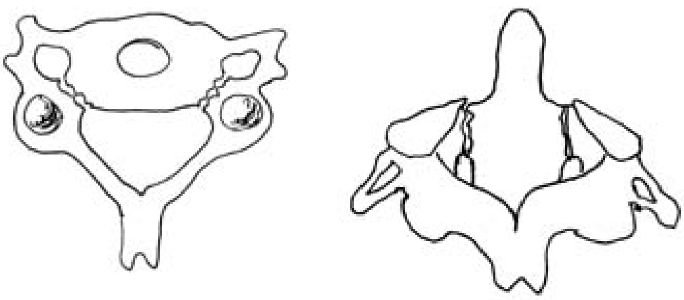


Figure 8: Hangman's fracture, superior and posterior view of C2 with fractured pedicles.

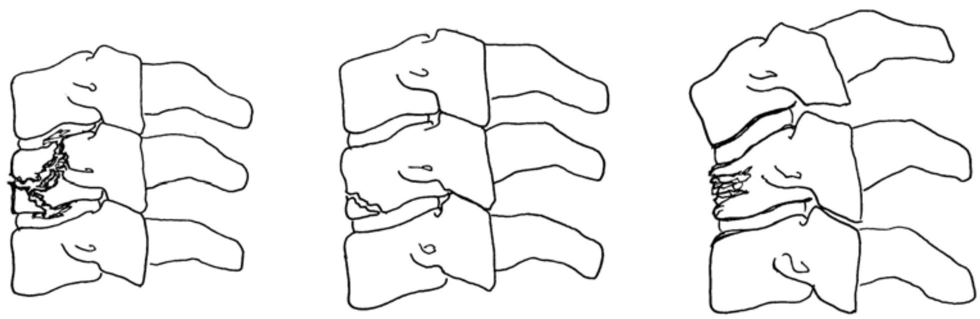
4.4 *Lower Cervical Spine Fractures*

4.4.1 Burst Fracture

In burst fractures the vertebral body disintegrates into several smaller fragments, Figure 9. This injury is frequently combined with fractures to the endplates and injury of the intervertebral disk. Burst fractures are often associated with neurological injuries, since the bony fragments can protrude into the spinal canal (An [1998], McElhaney and Myers [1993], Grane [1994], Wilcox et al [2002]). Critical loading for this fracture is thought to be severe axial compressive forces on the vertebral body. A numerical study by Bozic et al [1994] concluded that the fracture would begin in the central trabecular bone of the vertebral body. According to both An [1998] and McElhaney and Myers [1993] the most common fractures sites are C4-C5, C5-C6, and C6-C7. Burst fractures without neurological deficits can be treated with traction and external fixation, as the halo vest. For the cases where bone segments have injured the spinal cord surgical treatment, such as grafting, is necessary to enable recovery of the spinal cord functions (An [1998]).

4.4.2 Teardrop Fracture

The teardrop fracture is characterized by a triangular shaped bone segment that fractures from the inferior part of the vertebral body (on the anterior side), Figure 9. The injury is considered highly unstable in extension because the ALL is ruptured. The injury is stable in flexion since all the posterior ligaments are intact. According to An [1998] and McElhaney and Myers [1993] this injury is caused by flexion-compression loading, but according to Grane [1994] it can also be the result of local tensile loading of the ALL due to a global extension. External fixation is preferable for this injury if there are no neurological abnormalities found, which is suggested by among others An [1998] and Fujimura et al [1996].



A B C

Figure 9: Lower vertebral fractures. A) Burst fracture. B) Teardrop fracture. C) Wedge fracture.

4.4.3 Wedge Fracture

The wedge fracture is a failure of the anterior vertebral body, Figure 9C. This injury is thought to be the result of a flexion bending moment and a compression forces on the vertebral motion segment. According to McElhaney and Myers [1993] the most common sites are C4, C5, and C6. Carter et al [2000] concluded that slow compressive loading produced wedge fractures, while fast loading produced burst fractures. This fracture type is considered stable if the ligaments are intact.

4.4.4 Posterior Element Fracture

Fracture of the posterior elements of the cervical spine occur throughout the upper and lower cervical spine. This fracture can be isolated, but multiple fractures are frequent, Figure 10. These fractures include fractures of the laminae, the pedicles, the spinous processes, and the pars interarticularis. The bony structures that protect the spinal cord are injured. If this injury is associated with vertebral body displacement it is very unstable and requires internal fixation of several motion segments. The type of appropriate fixation depends on the fracture site and the patient's spinal cord status. This topic is discussed by among others An [1998]. According to McElhaney and Myers [1993] posterior element fracture is the result of bony contact between consecutive posterior elements, caused by extreme local extension. It is important to remember that local extension may appear in global extension, flexion, or compression of the neck.

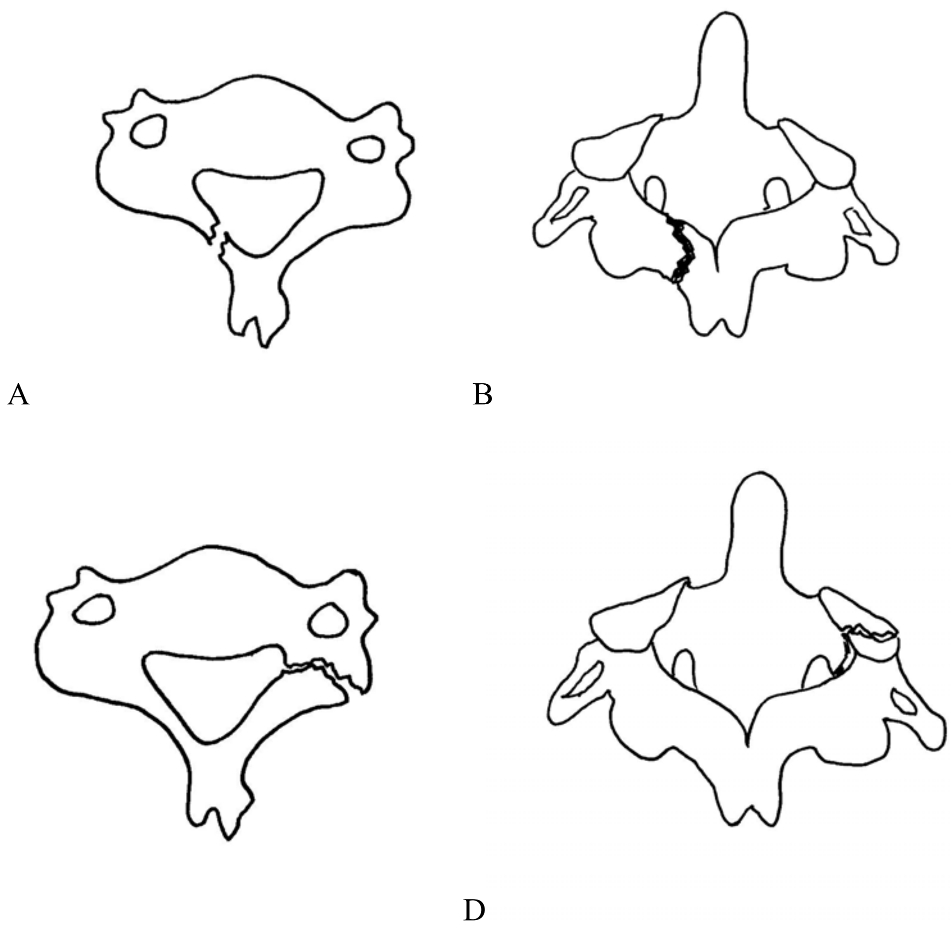


Figure 10: Posterior element fracture. Fracture of spinous process, superior view (A) and posterior view (B). Fracture of the lamina, superior view (C) and posterior view (D).

4.5 *Soft Tissue Injuries*

4.5.1 Facet Dislocation

In facet dislocations the superior vertebra is displaced anteriorly compared to the inferior vertebra, thus locking the facet joints, Figure 11. A dislocation can be either unilateral or bilateral, that is injury to only one or both of the facet joints. The bilateral facet dislocation causes a significant reduction of neural canal diameter, and is therefore often associated with spinal cord injuries. Neurological deficits are less common for the unilateral dislocations. The initial treatment is traction to reduce the dislocation, followed by a surgical treatment to stabilize the spine in axial rotation (An [1998]). It is common that facet dislocations are combined with facet fractures, thus complicating the injury and its treatment. According to An [1998] the loading mode producing this injury is local tension-flexion, while McElhaney and Myers [1993] argues that compression-flexion with tensile strains in the posterior ligaments is a more frequent failure mode. The same authors believe that unilateral dislocations are produced under the same loading conditions as the bilateral dislocations with the addition of local lateral bending or rotation.

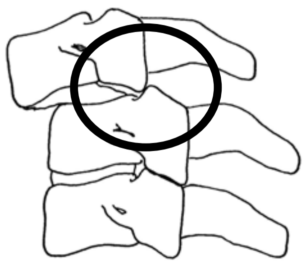


Figure 11: Facet dislocation between two vertebrae in the lower cervical spine.

4.5.2 Occipitoatlantal Dislocation

Dislocation at this level is the result of local tension in combination with other loading modes. This is a severe ligamentous injury often resulting in instantaneous death, Saeheng et al [2001]. The distraction of the spinal cord during the injury event is often associated with a lethal rupture of the spinal cord close to the brainstem. McElhaney and Myers [1993] and An [1998] agree that occipitoatlantal dislocations are rare, but that the true incidence is underestimated since most injuries are fatal. Those patients surviving usually have significant neurological deficits, showing symptoms such as quadriplegia. This injury requires immediate surgical treatment with internal fixation, to ensure spinal stability and to reduce spinal cord injury (An [1998]).

4.5.3 Atlantoaxial Subluxation

Atlantoaxial subluxation refers to injuries where one or both facet surfaces on C1 are displaced anterior or posterior to the facet surfaces on C2. Depending on the severity of displacement the alar, transverse, and capsular ligaments can be ruptured. An [1998] suggests that these injuries should be treated with traction until normal vertebral alignment is resumed. Then, either external or internal fixation is needed to stabilize the spine, for most patients. Atlantoaxial subluxation is produced by shear and torsion, or by a combination of both (An [1998], McElhaney and Myers [1993]).

4.5.4 Rupture of Ligaments

Ligament ruptures result from severe tensile strains in the ligaments. At first some of the collagen fibers break and if the tensile loading continues to increase the elastin fibers will start to fail, until complete failure of the ligament occurs. Tensile strains in anterior ligaments result from global extension motions, while posterior ligaments fail in global flexion. This has been studied experimentally by Maiman et al. [1983], who showed that spinal ligaments can easily be disrupted in flexion and extension. The load to failure varies for different ligaments and depends on size, proportion of collagen and elastin fibers, fiber organization and ligament position. Paper D lists reported failure loads of spinal ligaments in tensile testing. Ligament failure influences spinal kinematics differently. If several ligaments are ruptured, spinal instability can result as the restraining function of the ligaments are lost.

Alar ligament rupture mainly increases the motion range for axial rotation, but also allows for more relative motion between the atlas and axis, in some cases causing spinal cord compression. Transverse ligament rupture influences spinal stability, since the restraining band holding the odontoid process in contact with the anterior parts of the atlas is considerably weakened. ALL rupture can cause spinal instability if combined with other injuries. The same holds for the PLL, LF, and the SSL. CL rupture enables increased motion of the facet joints and can be combined with facet joint injury.

4.5.5 Subfailure of Ligaments

Lately, spinal researchers are hypothesizing that subfailure of the ligaments can be responsible for diffuse symptoms occurring in patients after low energy trauma, Panjabi [2002]. When the ligaments are stretched some of the collagen fibers may fail without any noticeable effect on spinal stiffness, this is defined as subfailure. Also, some of the neural cells apparent on the ligaments are injured. These cells will then send deficit information to the spinal muscles, causing them to react inappropriately. This in turn, could result in symptoms as headaches, back pain, dizziness, and etcetera.

4.5.6 Rupture of the Disc

Intervertebral disc trauma is frequently combined with vertebral body fractures (Maiman et al [1983]). Compression in combination with flexion or extension may cause this type of injury.

5 Numerical Models

The reader who is unfamiliar with numerical methods and especially the finite element method may want to consult page 30, where a short easy-to-read introduction is given, before proceeding with this chapter.

5.1 *Background*

The first numerical spinal models were developed in the late 1950's, to understand the biomechanics behind spinal injury due to pilot ejection. Since then, research with numerical techniques, such as the Finite Element Method (FEM), has made progress toward a better understanding of spinal behavior and injury mechanics. The FEM makes it possible to incorporate realistic geometry of the spinal tissues and their physical properties, to evaluate spinal kinematics, kinetics, and internal stresses and strains of the different spinal components. Lumped parameter models can also be used to study kinematics, but does not give the possibility to evaluate tissue stresses and strains since they are built up of rigid bodies. Knowing the stresses and strains in response to given load it is possible to predict the outcome, which is important for injury prevention. The anatomical details give a more reliable kinematical response than spherical representations of vertebrae and joints used in the lumped parameter models. Lumped parameter models of the cervical spine have been developed by Merill et al [1984], Deng and Goldsmith [1987], Li et al [1991], De Jager et al [1994, 1996], Camacho et al [1997, 1999], Winkelstein and Myers [2000], Nightingale et al [2000], among others. The first Finite Element (FE) spinal models were of the lumbar and thoracic regions; Belytschko et al [1973], Schultz et al [1973], Hakim and King [1979], Spilker [1980], Shirazi-Adl et al [1984, 1993, 1994], Kasra et al [1992], Lavaste et al [1992], Suwito et al [1992], Goel et al [1995], Sharma et al [1995], Kiefer et al [1998], Whyne et al [1998], Ladd [1998], to mention a few.

5.2 *Single Cervical Vertebral Models*

Cervical FE models of single vertebrae have been developed from CT-images and used to analyze stress patterns in the vertebral bone during loading, in an effort to predict fracture mechanisms. Teo et al [1994] modeled C2 with anatomical detail but assumed the vertebra to be entirely made up of cortical bone, while Graham et al [2000] included both cortical and trabecular bone in their analysis of odontoid fractures. Teo and Ng [2001a] and Bozkus et al [2001] studied fracture patterns in two different anatomically detailed C1, though idealized the vertebra as only composed of compact bone. In a model of C4 by Bozic et al [1994] the elastic modulus was calculated independently for each element, although the strict cubical shape of the elements decreased anatomical detail.

5.3 *Cervical Motion Segment Models*

FE models of cervical motion segments have been developed to study load distribution between the vertebrae and biomechanical effects secondary to impact; Clausen et al [1997], Maurel et al [1997], Yoganandan et al [1997], Ng and Teo [2001], Teo and Ng [2001b], Wilcox et al [2002]. All the mentioned cervical models had linear elastic material characteristics. Kumaresan et al [1997c] developed an anatomically detailed C4-C6 motion segment of the adult human and three age-specific one, three, and six year old models. The vertebrae were represented with shell elements for the cortical bone and solid elements for the trabecular bone. The inter vertebral disc was modeled with annulus fibers and ground substance in combination with linear elastic solid elements for the nucleus pulposus, which is the first model to include anisotropic behavior of the disc. Non-linear spring elements were used to model the ligaments. Validation was performed in quasi-static compression, flexion, and torsion. In a parameter study of the material properties in the lower cervical spine Kumaresan et al [1999] separately varied the different properties of the cervical tissues. They found a change of Young's modulus for the hard tissues to affect the response of the hard tissues, while a change of soft tissue properties affect the kinematics as well as the response of the soft and hard tissues. Wheeldon et al [2000] added C2, C7 and T1 to the C4-C6 model by Kumaresan [1999], thus having a model of the complete C2-T1 segment. Validation of this model has not been published. Goel and Clausen [1998] implemented non-linear material properties for the ligaments in an FE model of the C5-C6 complex. Validation was conducted for compression, flexion, extension, axial rotation, and lateral bending.

5.4 *Models of Cervical Injury*

Cervical spinal models were developed by several authors to study fracture fixation methods in the cervical spine, Kumaresan et al [1997a, 1997b], Voo et al [1997], Nataranjan et al [2000], Lim et al [2001], Puttlitz et al [2001], Pitzen et al [2002].

5.5 *Complete Cervical Spinal Models*

The first model of the complete cervical spine was developed by Kleinberger [1993]. It included a rigid head and was made for dynamic simulations. The vertebrae were rigid bodies, the intervertebral disc was homogenous, and the ligaments were modeled with brick elements. All soft tissues were given isotropic, linear elastic material properties. The upper cervical spine was modeled as separate vertebrae and included the odontoid process on the axis. The occipitoatlantal joint was simplified as a pin joint, allowing for rotation about one axis (flexion and extension motion) and maintaining atlas/skull contact in tension. Kleinberger's model was validated for axial compression and frontal impacts. Recently, Stemper et al [2000] varied the material properties in Kleinberger's model during a frontal impulse applied at T1. They found that the rotation of the center of gravity for the head increased when the elastic moduli of the cervical vertebrae and discs were increased, while the Poisson's ratio did not affect the head motion. Dauvilliers et al [1994] created another FE model of the cervical spine and included a rigid head, also for dynamic simulations. They also modeled the vertebrae as rigid bodies, while the intervertebral disc was modeled with homogenous brick elements, and the ligaments with tension-only springs. All material properties were linear elastic, although damping was included for the ligaments. The lower cervical spine was modeled with anatomical detail while the upper cervical spine was simplified. The occipitoatlantal and atlantoaxial joints were modeled as two spherical joints, thus allowing for flexion-extension motion and axial torsion between the skull and the vertebral body of C2. This model was validated for frontal and lateral impacts. Another head and neck model was developed by Yang et al [1998]. The vertebrae were given an elasto-plastic constitutive law, although the cortical and trabecular bone was simplified and assumed to have the same stiffness. The intervertebral disc was divided into a linear viscoelastic model of the nucleus pulposus and linear elastic elements for the annulus fibrosus. The ligaments were represented with nonlinear tension-only membrane and bar elements. All ligaments in the same location but at different spinal levels were assumed to have the same elastic modulus. Data for the ALL from the lumbar spine was used and scaled to represent the ALL, PLL, SLL, ISL and LF in the cervical spine. The upper cervical spine was modeled anatomically and free sliding was defined between all contact surfaces. This model by Yang et al was compared to experimental head-neck drop tests and rear impact sled tests. Jost and Nurick [2000] developed another complete cervical spine model with a rigid head, as a part of an FE model of the whole body.

The vertebrae were modeled with shell elements to simulate the cortical bone. The trabecular bone in the vertebrae was neglected. Ligaments were modeled with tensile membrane elements with linear elastic material properties, in combination with spring-damper elements to model the viscoelastic material behavior. The lower and upper cervical spine was modeled with anatomical detail. Thus, this is the first reported model of an anatomically correct upper cervical spine with both compact and trabecular bone. The model was validated at low and high g-level lateral accelerations and only the resulting head motion was evaluated. At the moment, the Defense Science and Technology Agency in Singapore is developing a complete cervical spinal model of C1 to C7, which has not been published yet, Seng [2002]. This model is based on static FE analysis.

There is still much work to be done before a validated FE model of the complete cervical spine is available. More complex material models need to be developed to simulate biological materials, but there is also the need for relevant data on material properties of cervical tissues.

5.6 The Upper Cervical Spine

Another important feature that FE models needs is to implement the coupling between the head and the neck. All models, except the ones by Yang et al [1998] and Jost and Nurick [2000], have simplified the joints and the anatomy of the upper cervical spine. The model by Jost and Nurick includes an anatomically accurate upper cervical spine, the occipitoatlantal and atlantoaxial joints are modeled with facet surfaces, where the space between the surfaces was filled with linear elastic solid elements and a system of springs and dampers, as in the model by Dauvilliers et al [1994]. Yang et al [1998] defined the upper cervical joints as free sliding between bony contact surfaces, and their model did not include a separation of cortical and trabecular bone. Jost and Nurick's model was developed for lateral impacts and Yang's model was also validated globally.

There is a need to model the upper cervical spine with anatomical detail, differentiate between the cervical tissues and to validate the upper cervical spine locally as well as globally, together with the lower cervical spine and the head.

6 Material and Method

6.1 *Statistical Survey*

Two national studies of statistical data in Sweden were conducted as a part of this thesis, Paper A and B.

The statistics in Paper A and B are defined by descriptive analysis. The Swedish Hospital Discharge Register (HDR) was used as a basis of information for this study. Statistics of diseases and surgical treatment of patients have a long history in Sweden. Data of this kind have been published for more than 100 years and have been available for the whole 20:th century. In the 1960:s the National Board of Health and Welfare started to collect data on individual patients, who had been treated as in-patients at public hospitals. The register was built up at that time and in 1983 twenty of the 26 county councils reported all in-patient care to the HDR. In 1984 reporting to HDR was made compulsory.

These studies are based on data from 1987 and from this time the HDR includes all public, in-patient care in Sweden, while all out-patient data from general practitioners are excluded. Today, there are 93 emergency hospitals reporting to HDR. The information to HDR is delivered once a year to the Center for Epidemiology (EpC), at the National Board of Health and Welfare in Sweden, from each of the 26 county councils in Sweden. There are four different types of information in the HDR: data on the patient, data on the hospital, administrative data, and medical data. Patient data includes personal identification number (PIN), gender, age, and place of residence. Administrative data includes date of admission and discharge, and length of stay. Medical data includes main diagnosis, external cause of injury (E-code), surgery, medical treatment, and fatality.

The reliability of the data from the HDR is well controlled and the underreporting in HDR has been estimated by using statistics on hospital stays, from the Federation of County Councils in Sweden. The total number of dropouts for somatic short-time care is estimated to less than two per cent. Quality insurance is performed routinely on data reported by each county and hospital. Some obviously incorrect data is corrected in connection with the quality controls. As an example, in 1996, the number of stays with missing PIN was 0.95 percent, and the main diagnosis was missing in 1.46 percent of the hospital stays reported. The number of hospital stays with an injury diagnosis, where the E-code was missing, has increased from 3.05 percent in 1987 to 3.75 percent in 1996. One county council is responsible for two thirds of the missing E-codes.

Data between 1987 and 1996 were registered according to the International Classification of Diagnosis revision 9 (ICD 9) and data from 1997 to 1999 on revision 10 (ICD 10). However, a few counties still reported according to ICD 9 in 1997. In ICD 9 the two main diagnoses were fractures of the cervical vertebrae with and without spinal cord injury (805 – 806) while the main diagnoses in ICD 10 distinguishes between the locations of the vertebral fractures (S12.0 – S12.9) and also includes soft tissue injuries in the cervical spine (S13.0 – S13.6). Therefore, studies of the soft tissue injuries and detailed studies of fracture type were only possible from 1997 to 1999. Data on fatal injuries are reported for the time period between 1987 and 1998.

6.2 *Finite Element Model*

An FE model of the human head and neck was developed jointly with Peter Halldin and Svein Kleiven, Figure 12. Figure 13 shows the FE model of the upper cervical spine that was developed as a part of this thesis, to give an adequate coupling between the head model (Kleiven [2002]) and the neck model (Halldin [2001]). Papers C, D, and E describe the FE model in more detail, with focus on the upper cervical spine. This chapter gives a short introduction to the FE model of the upper cervical spine, which includes the occiput to the third cervical vertebra, C3. LS-DYNA was used for all FE calculations, Hallquist [1998].

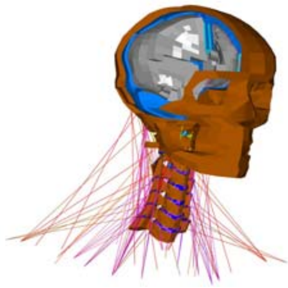


Figure 12: The complete head, brain and neck FE model developed together with Ph.D. Peter Halldin and Ph.D. Svein Kleiven.

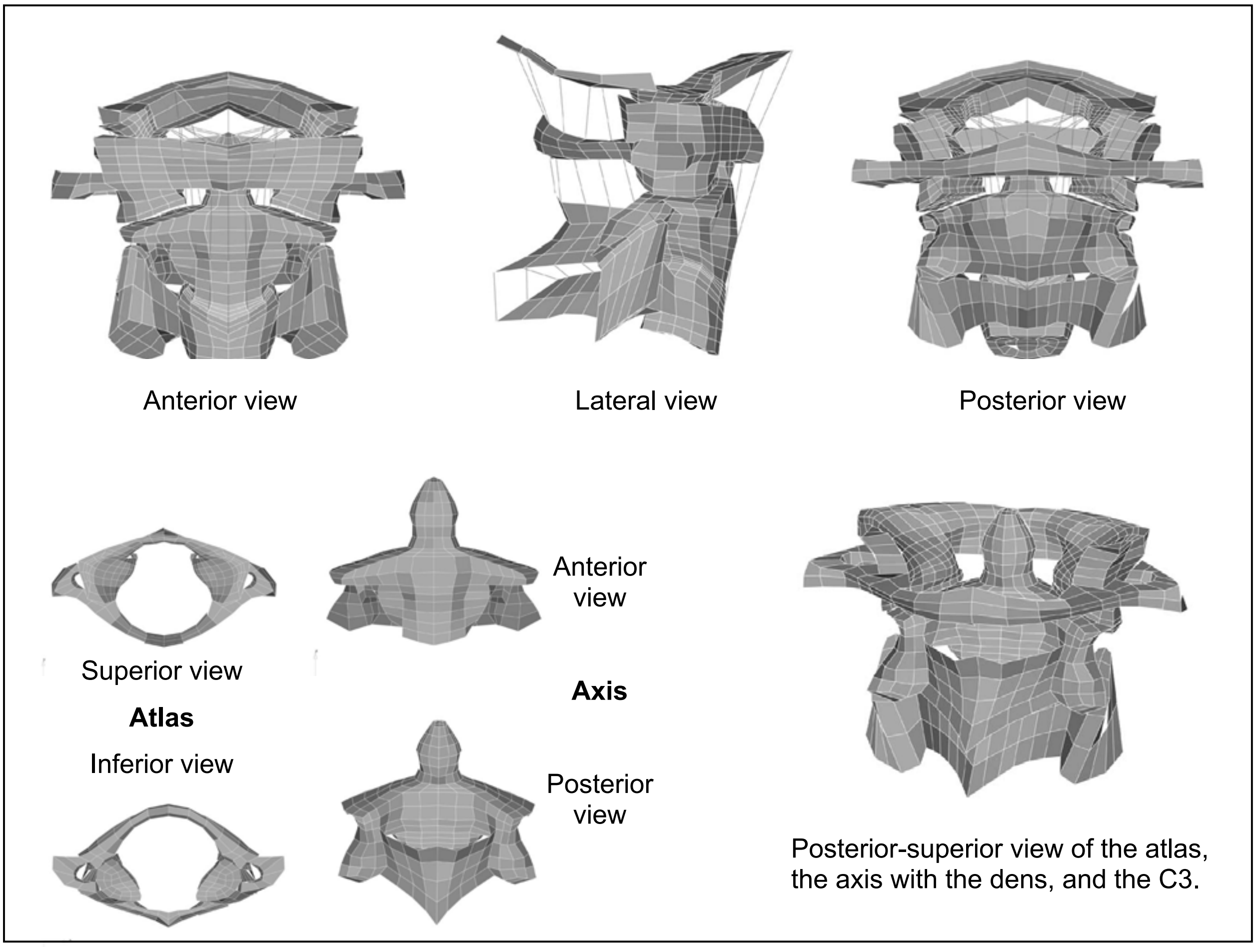


Figure 13: The FE-model of the upper cervical spine. The top row shows the C0-C3 model in anterior, lateral, and posterior views. In the bottom row the C1, C2, and C1-C3 segment is shown.

6.2.1 Geometrical representation

An anatomically detailed model gives several advantages such as correct inertial properties and well defined contact surfaces between the vertebrae, resulting in well-defined motion.

CT images of a 27-year-old man were used to determine the exact dimensions of the bony areas of the cervical spine. The images were equally distributed with 1 mm distance through the spine. Slides were taken in the sagittal and coronal planes. The coordinates from each CT slide were transferred, using the software NIH Image 1.95 software, to an FE code pre processor called ALADDIN, OASIS-ALADDIN [1993]. The transferred points were then used to build surfaces representing the cortical bone, and solids representing the trabecular bone. All surfaces were meshed with 4-node plane elements and the solids with 8-node 3D elements. Later refinements and model changes were done with another pre processor, Hypermesh (from Altair Engineering, Inc, www.altair.com).

Ligaments were added to the vertebral model. The orientation and length of spinal ligaments were found in the literature; Panjabi et al [1980, 1991c], Myklebust et al [1988], Goel et al [1988], Dvorack et al [1988], Saldinger et al [1990], Crisco III et al [1991a, 1991b], Przybylski [1998]. The TL was modeled with membrane elements, to account for contact with the odontoid process, Figure 14. All other ligaments were discretized with spring elements: the ALL, the AAOM, the PLL, the CL, the TM, the LF, the PAOM, the alar ligament, the apical ligament, the VC, the ISL and SSL, Figure 14 and Figure 15.

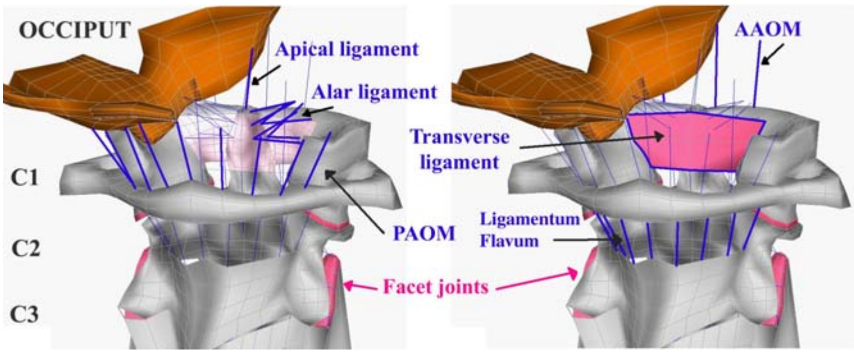


Figure 14: Posterior view of the upper cervical spine, with some of the ligamentous structures highlighted.

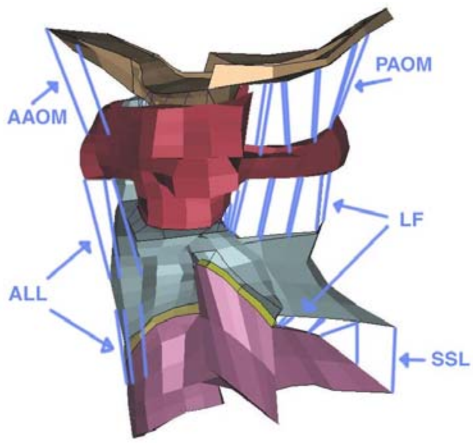


Figure 15: Lateral view of the upper cervical spine with some of the ligaments highlighted.

The cervical joints were included in the model. The occipitoatlantal joint, atlantoaxial joint, and facet joints were modeled with sliding contact surfaces. To account for the response of cartilage, a layer of solid elements was added to one of the joint surfaces. Contact was then defined between the solid elements and the second joint surface. The occipitoatlantal joint does not include these solid elements. Capsular ligaments were added as spring elements to all the synovial joints. A detailed model of the intervertebral disc between C2 and C3 was used, Halldin [2001]. The annulus fibrosus was represented with three orthotropical cylindrical layers, each layer composed of five lamellae. The nucleus pulposus was modeled by solid elements.

6.2.2 Material properties

Biologic materials are anisotropic, inhomogeneous, non-linear, time-dependent, and viscoelastic. This complicates the modeling of biologic tissue, compared with traditional engineering materials. Another difficulty, that arises when human tissue is used, is the great span of values between individuals. Therefore, all the material properties are approximated, and can, compared with a specific specimen or individual, seem far off. There is, indeed, a lack of experimental data for living human tissue, which demands simplifications to less complicated material models.

The vertebrae are modeled with isotropic, linear elastic material models. The cortical bone was given a Young's modulus of 15 GPa. Young's modulus for the trabecular bone in the C1 and C3 vertebrae was 0.5 GPa and in the C2 vertebra 0.75 GPa. These values were derived from the pixel value of the CT images. According to Carter and Hayes [1977], there is a relation between the apparent density (ρ_a), the strain rate ($\dot{\varepsilon}$), and Young's modulus (E): $E = 3790\dot{\varepsilon}^{0.06}\rho_a^3$. Here a strain rate of 0.1 s^{-1} was assumed. The apparent density is proportional to the pixel value and was calculated to be 1.8 g/cm^3 and 0.2-0.5 g/cm^3 for cortical and trabecular bone respectively. The range of experimental Young's modulus is 15 – 21 GPa for cortical bone and 0.1 – 5 GPa for trabecular bone (Hakim and King [1979], Halldin [2001]). Poison's ratio of 0.2 was assumed for both cortical and trabecular bone.

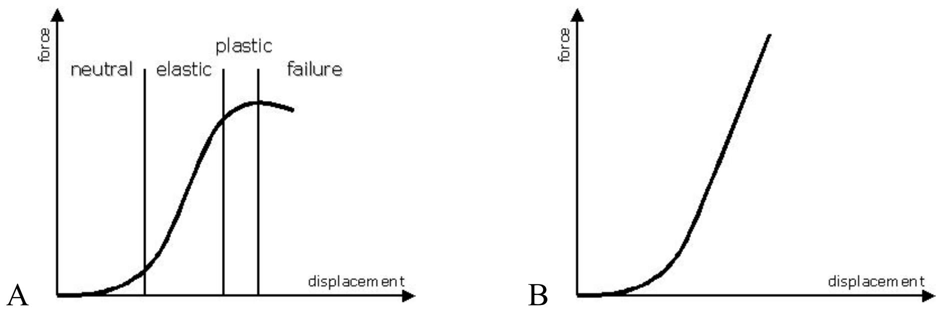


Figure 16: A) Force-displacement curves for ligaments from mechanical testing with the characteristic neutral, elastic, plastic, and failure zones. (The curve is redrawn from White and Panjabi [1990]) B) Force-displacement curve used to define material properties of ligaments in the FE model of the upper cervical spine.

The ligaments were modeled with linear tension-only springs in Paper C, and then improved by implementing the non-linear material properties in Papers D and E. The proportional content of elastin and collagen and the structural orientation influences the mechanical properties of ligaments. The force-displacement curve of ligaments typically has a flat toe region, linear physiological phase, a traumatic phase, and then a post-traumatic phase before the ultimate load carrying capacity of the structure is reached, (Yoganandan et al [1989b, 2001], Crawford et al [1998], White and Panjabi [1990]). These different regions are also called the neutral, elastic, plastic, and failure zones, Figure 16A. In this study, the ligaments were assigned force-deflection curves containing the neutral and elastic zones, thus failure was not implemented in this model. The length of the toe region was initially assumed to be one third of the failure deformation and the elastic zone was continued linearly past the point of ligament failure, Figure 16B. Force and deflection at failure for the different ligaments were found in the literature (Myklebust et al [1988], Dvorack et al [1988], Yoganandan et al [1989b]). The time-dependent response of spinal ligaments was neglected, and data from quasi-static experimental tests used. The force-deflection curves used for each ligament are listed in Paper C.

The material properties of the intervertebral disc were taken from Halldin [2001]. The nucleus pulposus was a Poisson's ratio of 0.4999, thus simulating the incompressible nature. The annulus fibrosus was divided into the annulus ground substance and an orthotropical composite layer representing the annulus fibers. Paper C and D list all material properties.

Viscoelastic material models have been implemented for all cervical tissues, Emanuel [2001], although these data were not used in the appended papers.

6.2.3 Neck musculature

Neck musculature has been added to the FE model, mainly through the joint efforts of Ingrid Leijonhufvud and Peter Halldin, Figure 17. This work started as a master thesis project by Leijonhufvud [2001] and is still ongoing. Leijonhufvud modeled the muscles groups with two-node spring and damper elements, based on anatomical data in the literature. This model has been improved by introducing multi-node spring elements to enable muscle curvature around the vertebral column, Table 1. In the work by Leijonhufvud only non-linear passive muscle properties were implemented, recently FE studies of active muscle properties in flexion have been conducted.

Table 1: List of muscles modeled with two- and multi-node springs.

Musclegroups	nof springs	multi-node
Sterno Cleido Mastoid	24	yes
Longus Cervicis	19	
Rectus Capitis Anterior Major	8	
Rectus Capitis Anterior Minor	4	
Scalenus anterior, medius, posterior	24	
Suboccipital Muscles	8	
Semispinalis Capitis, Semispinalis Cervicis	40	yes
Longissimus Capitis	62	yes
Splenius Capitis, Splenius Cervicis	40	yes
Levator Scapulae	32	yes
Trapezius	66	yes
Interspinous muscles	5	

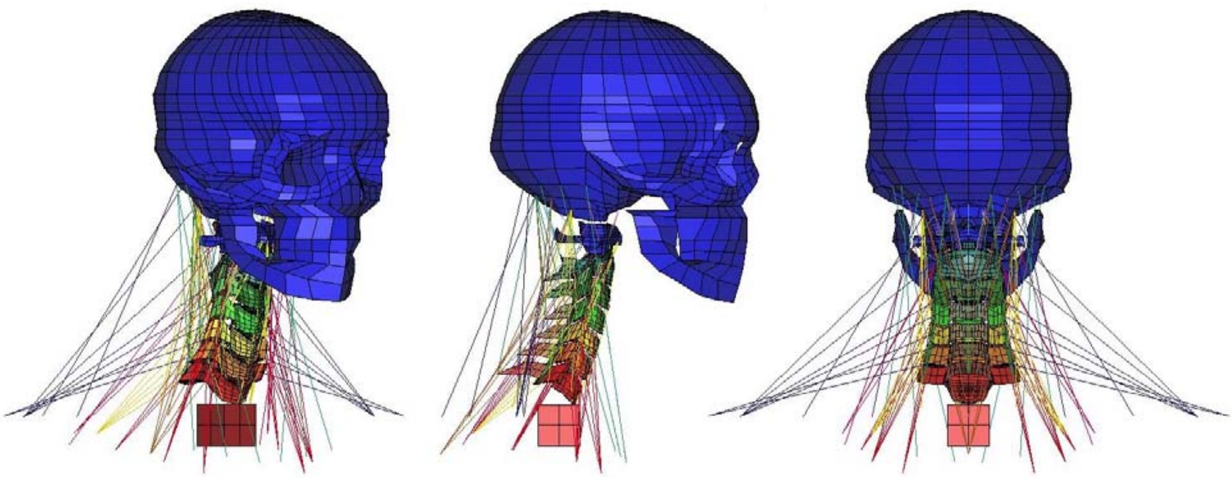


Figure 17: The FE neck muscle model developed by Leijonhufvud [2001].

6.2.4 Comparison with experiments

It is important to verify that a numerical model exhibits a realistic response. The FE model of the spine was compared to experimental data for many different load cases, Table 2. Paper C describes some of the validations for the complete neck model for dynamic loading and Halldin [2001] has validated the lower cervical spine separately. Paper D studies the angular responses of the three upper cervical joints, as a function of the applied loads. Those validations were for quasi-static flexion, extension, axial rotation, lateral bending, and tension. The head and neck model with passive muscles have been validated against volunteer sled data, Leijonhufvud [2001]. Unpublished data confirm correlation of the active muscle model to the same data.

Table 2: Conducted validations of the FE neck model up to August 2002.

	Anatomical region	Loading mode	Type of comparison	Correlation	Reference of experiments	Published results
Local validation	C4 - C5	compression	load/deflection	Good	Liu et al [1980]	Halldin [2001]
		shear	load/deflection	Good		
		torsion	load/deflection	Good		
	C4 - C5	compression	injury prediction (disc herniation)	Realistic	Halldin [1998]	Halldin [2001]
	Occiput - C3	flexion	motion	Good	Panjabi et al [1991a,b]	Paper C, D
		extension	motion	Very good		
		axial rotation	motion	Good		
		lateral bending	motion	Very good		
	Occiput - C3	tension	load/deflection	Very good	Van Ee et al [2000]	Paper D
	Occiput - C3	axial rotation	load/deflection	Good	Goel et al [1990]	Paper D
Complete ligamentous cervical spine	Occiput - C7	compression	motion	Very good	Nightingale et al [1996, 1997]	Paper C
	Occiput - C7	compression	boundary conditions	Good	Nightingale et al [1996, 1997]	Paper C
Complete cervical spine with simple muscle model	Occiput - C7	frontal impact	motion	Good	Ewing et al [1976]	Leijonhufvud [2001]
			acceleration	Realistic, but oscillatory		
	Occiput - C7	lateral impact	motion	Good	Ewing et al [1976]	Leijonhufvud [2001]
			acceleration	Realistic, but oscillatory		
	Occiput - C7	oblique impact:	motion	Good	Ewing et al [1976]	Leijonhufvud [2001]
			acceleration	Realistic, but oscillatory		

7 Discussion

In order to assess the consequences of injuries to the public health and to develop and implement effective injury prevention programs, it is necessary to describe the magnitude, populations at risk, etiologic factors as well as severity and outcome. Studies of epidemiological data are an important means of defining the scope of the problem. Papers A and B in this thesis, provide an overview of neck injuries in Sweden. To the best knowledge of the author, Paper A is the first published national neck incidence study. It was shown that Swedish men are more likely to sustain neck fractures than women. This is true for all age groups. Hereby, neck injuries differ from other common fractures, as hip or femoral fractures, where elderly women are clearly over represented, Boström [2001]. For the elderly population, all these different fractures are mainly caused by fall injuries. In contrast, the younger population often injures the neck in traffic accidents. Though, it should be noted that when an individual falls the risk of sustaining a neck injury is not dependent on age. Therefore, it can be assumed that neck injuries in the elderly population may be reduced by prevention of falls. Also, Paper A illustrates how fall accidents account the majority of the cervical injuries.

Although, traffic accidents related neck injuries have decreased over the years there is still a need for preventive efforts. One restriction of the data in Paper A is the exclusion of accidents that are immediately fatal. The incidences of neck fractures at high locations are underestimated, for this reason, Saeheng [2001. Therefore, there is a need for novel safety systems like the inner car roof developed in Paper C. Today, the lack of detailed injury thresholds for the human limits the preventive research, in for example the vehicle industry. The FE model developed in this thesis can speed the progress of prevention by introducing tissue level criteria. This was proved to be true since the inner roof safety system in Paper C, which could not have been designed or evaluated only with existing injury criteria, was developed using the FE model. It is very likely that the role of human FE models will be as large as crash simulation is today in the car industry.

A striking result from Paper A is that risk for upper cervical fractures is almost twelve times larger for the population over 65 years of age than for the younger age groups. Other studies have reported similar results, Spivak [1994] and Lomoschitz [2002]. The latter proposed that a stiffened cervical spine due to normal aging causes the higher incidence for the elderly. Paper B also states that the length of hospital treatment is somewhat longer for the elderly population. Paper D illustrates how important the ligaments in the upper cervical spine are to maintain spinal stability. Spinal instability in the upper cervical spine may have fatal consequences, as this is where the nerve roots to vital organs leave the spinal cord. In order to give the patient with a high neck injury an optimal treatment it is necessary to have a thorough understanding of the structural effects of the injury, which reflects the degree of instability. The results from the study of ligamentous ruptures in Paper E provide evidence of the complexity of the upper cervical spine. From that data it can be concluded that the FE model can contribute to the study of this complex system. It is tentative to proceed with studies of how vertebral fractures are affected by simultaneous soft tissue injuries and with studies of upper cervical injury fixations methods.

The FE model has some advantages compared to volunteer and experimental studies. It is possible to study the stresses and strains in individual tissues with good control of boundary and loading conditions, which is much more difficult in volunteer experiments. In the FE model complex injuries can be implemented, while cadaver experiments are limited since the outer tissue will inevitably be injured before the more central tissues can be transected. If injuries are mechanically initiated in experiments the outcome is always somewhat unpredictable, whereas the exact location of an injury can be precisely controlled in an FE model. On the other hand, the FE model will never be better than the data put in to it. Therefore, excessive validation is a major issue for FE modeling. Paper E revealed some limitations of the FE model developed in this thesis, that will have to be resolved before more detailed studies can be conducted. The strength of this FE model is that it combines the head and the neck. The interaction between the head and the neck highly influences injury mechanisms as well as the stability behavior. Therefore, a model of the whole complex will enable reliable studies of neck issues.

8 Conclusion

It is concluded that neck injuries in Sweden is a problem that needs to be addressed with new preventive strategies. It is especially important that results from the research on fall accidents among the elderly are implemented in preventive programs.

Secondly, it is concluded that an FE model of the cervical region is a powerful tool for development and evaluation of preventive systems. Such models will be important in defining preventive strategies for the future.

Lastly, it is concluded that the FE model of the cervical spine can increase the biomechanical understanding of the spine and contribute in analyses of spinal stability.

9 Future Work

Future research concerning the FE model should focus on the following aspects:

- Remodel the ligaments with membrane elements that can account for ligament-ligament and ligament-vertebra contacts.
- Complement the current muscle model with solid elements that represent the volume effect of the spinal muscles.
- Model the superior part of the thoracic spine, the clavicle, and the scapula to enable more accurate insertion of the cervical muscles and to apply more realistic boundary conditions.
- Perform further validations as soon as new experimental data are available.

The medical applications are a very important field to explore in future research with FE models. Following an impact to the cervical spine, the patient is the object for different imaging technologies, such as CT and MRT, in an effort to evaluate the injury. The aim is to find the optimal treatment for each patient. With the introduction of an advanced FE model of the neck, it is tentative to speculate that it might be possible to evaluate data of importance for improvements of individual treatments.

Also, better criteria are need for the progress in primary prevention. The FE model will be used to scale current neck injury criteria to other injury severity levels. Tissue failure criteria applied in an FE model can be used to determine new neck injury criteria. These criteria can be adjusted to combined loading conditions and should be applicable to output from for example car crash dummies.

Terminology

Directions, Figure 18

Anterior	Towards the front
Inferior	Below
Lateral	Towards the side
Posterior	Towards the back
Superior	Above

Anatomical Planes

Coronal plane
Figure 18
Sagittal plane
Figure 18
Axial plane
Figure 18

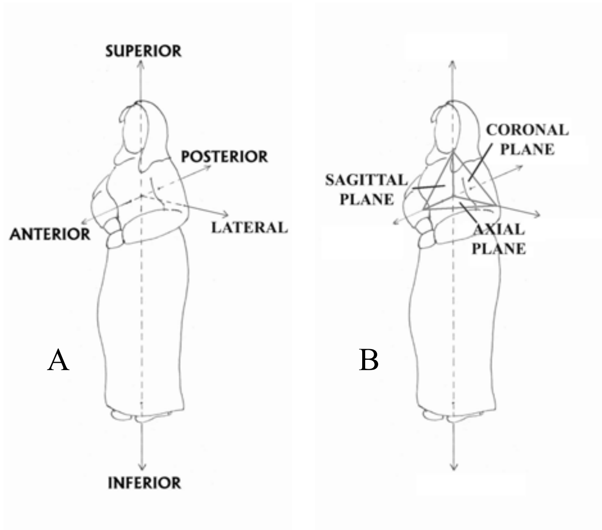


Figure 18: Directions (A) and anatomical planes (B).

Motions

Axial Rotation	Figure 5, page 9
Extension	Figure 5, page 9
Flexion	Figure 5, page 9
Lateral Bending	Figure 5, page 9

Anatomy

AAOM	Anterior atlantooccipital membrane, section 3.3
AIS	Abbreviated Injury Scaling, section 4
Alar ligament	section 3.3
ALL	Anterior longitudinal ligament, section 3.3
Annulus fibrosus	The outer thick structure of the disc
Apical ligament	section 3.3
Atlantoaxial joint	Joint between the first and second cervical vertebrae, section 3.2
Atlas	First cervical vertebra, C1, Figure 2
Autonomous system	Nervous system for heart, breathing, and other vital functions
Axis	Second cervical vertebra, C2, Figure 2
Bilateral	On both sides
C1, C2, … C7	Cervical vertebra number 1, 2, … 7
C1/C2, etc	Motion segment including cervical vertebrae 1 and 2.
Cervical Spine	The superior region of the spine with the 7 vertebrae
CL	Capsular ligament, section
Coccyx	A small triangular bone at the base of the spine, Figure 1
Cortical bone	Compact bone at the surface of the vertebra, section 3.1
Craniocervical junction	The occipitoatlantal joint, section 3.2
CT	Computer Tomography
Dens	Part of the second cervical vertebra, section 3.1
Endplate	Stiff, bony connection between the intervertebral disc and the vertebrae
Facet joint	Synovial joint in the spine, section 3.2
Fibro cartilaginous joint	Joint composed of fibrous tissue that only allows small motions
Grafting	Replace an injured vertebral body with another piece of bone or implant material and fixate the adjacent vertebrae, section 4
Halo Vest	A devise used to stabilize an injured cervical spine. It locks the shoulder region with the head through an external ring and screws in the skull bone, section 4.

Intervertebral disc	Soft tissue joint in the spine, connecting two adjacent vertebrae, section 3.2.
ISL	Interspinous ligament, section 3.3
Lamina	Part of a vertebra, section 3.1
LF	Ligamentum flavum, section 3.3
Ligaments	Restraining bands in the spine, section 3.3
Lower Cervical Spine	The cervical spine from vertebra three to seven, Figure 2
Lumbar spine	Spinal region below the thoracic spine
MRT	Magnetic Resonance Tomography
Neural canal	Spinal canal
Nucleus pulposus	The inner gelatinous structure of the disc
Occipital condyles	Part of the occiput, section 3.2
Occipitoatlantal joint	Joint between the occiput and the first cervical vertebra, C1, section 3.2
Occiput	Posterior part of the head
Odontoid process	See Dens
PAOM	Posterior atlantooccipital membrane, section 3.3
Paraplegia	Bilateral paralysis, usually the lower part of the body
Pars Interarticularis	The part of the vertebra between the facet surfaces, Figure 2
Pedicles	Part of the posterior arch of a vertebra, Figure 2
PLL	Posterior longitudinal ligament, section 3.3
Posterior arch	Part of a vertebra, Figure 2
Sacrum	A large wedge shaped bone in the lower part of the back, below the lumbar spine, Figure 1
Spinal Cord	Thick cord of nerve tissue within the spine, which connects the brain and the peripheral nerves in the body
Spinous process	Part of the posterior arch of a vertebra, Figure 2
SSL	Supraspinous ligament, section 3.3
Stable injury	An injury for which the spine is still stable.
Stable spine	The spine is able to protect the spinal cord during normal physiologic loading, section 4
Synovial joint	A bone joint with two contact surface and a joint capsule. The bone surface is covered with cartilage, thereby creating a low friction joint.
T1	First thoracic vertebra, Figure 1
Thoracic spine	The spinal region below the cervical spine, Figure 1
TM	Tectorial membrane, section 3.3
Trabecular bone	Porous bone in the center of the vertebrae, section 3.1
Traction	Treatment where patients neck is straightened by attaching weights to the head, section 4
Transverse ligament	Restrains the dens to the atlas, section 3.3
Transverse process	Part of the vertebra, Figure 2
Unstable injury	An injury that will makes the spine unstable
Unstable spine	The spine is unable to protect the spinal cord during normal physiologic loading, section 4
Upper cervical spine	The cervical spine from the first to the third vertebrae, Figure 2
VC	Vertical cruciate, section 3.3
Vertebra	Bony segment in the spinal column, section 3.1

A Brief Introduction to Finite Element Analysis

This section briefly explains the Finite Element (FE) technique and is intended for those who have never come across FE analysis before nor have the relevant technical background. The interested can find more detailed information in one of the many books written on the subject (among others Cook et al, Szabó and Babuška, and Bathe).

FE analysis is a numerical method, which means that it simplifies the real world using mathematical formulas. FE procedures are today an important and valuable part of engineering analyses and design of complex structures. Figure 19 illustrates the process of an FE analysis. The structure to be studied is divided into a finite number of elements, connected at nodes. Then, a set of algebraic equations defining the problem is established and solved for all elements and nodes, using computer resources. The results can, for example, be displayed as displacements, stresses or strains in the structure throughout the load duration. Several simplifications are made in an FE analysis. First, the actual geometry is simplified with 1-D, 2-D, or 3-D elements. Secondly, the mathematical representation of the material properties, and other features of the problem, are simply models and can be chosen with various degrees of complexity. Therefore, it is important to verify or validate that the response of the model is realistic when the results are analyzed.

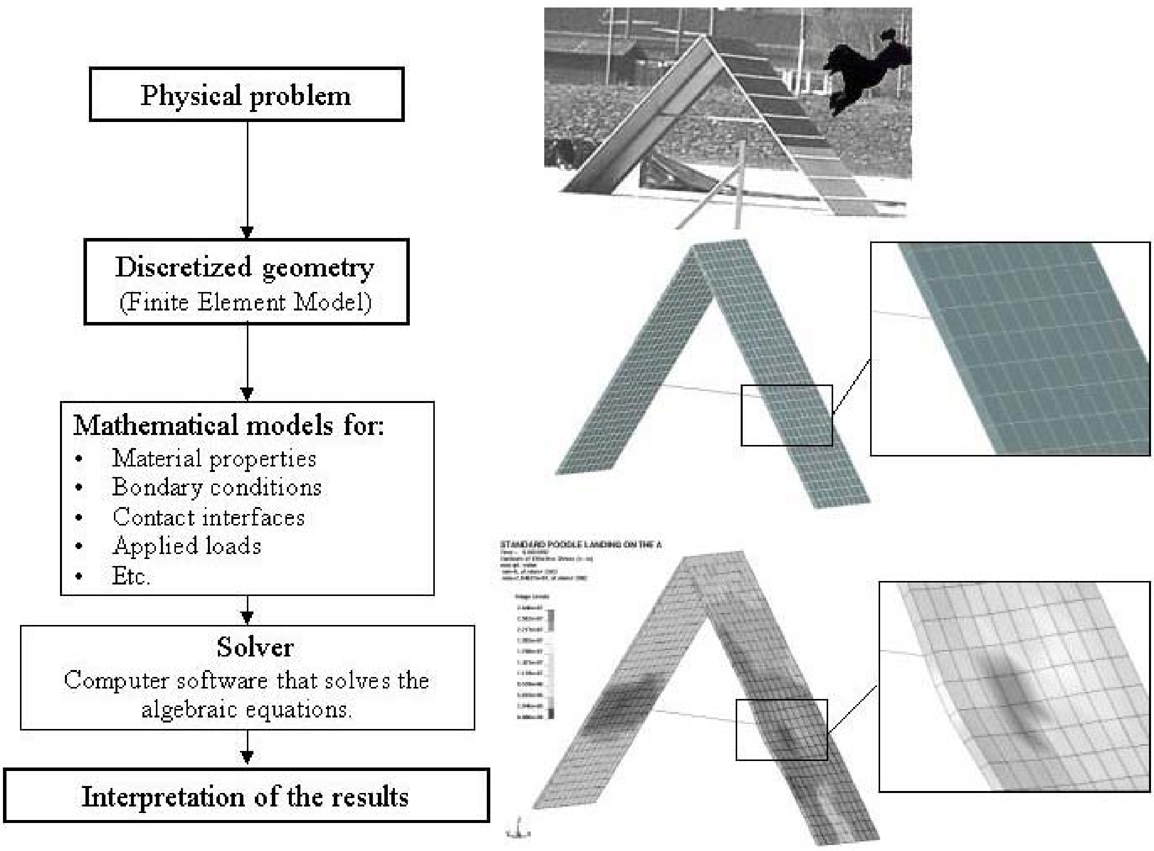


Figure 19: Schematic illustration of the Finite Element techniques illustrated with one example of a dog landing on a barrier. The top image illustrates the physical problem, the second image displays the FE model with elements and nodes at the intersections, the bottom image shows the resulting deformation calculated by the solver. The darker colored patches, representing elements where the stress is highest, show that the elements close to the point of contact as well as some elements on the opposite side, near the metal rod, take most of the loads.

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