Mathematical modelling of the human cervical spine

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Mathematical Modelling of the Human Cervical Spine: A Survey of the Literature

Marko de Jager

Abstract

Injuries to the cervical spine are quite commonly found in traffic accidents. The mechanisms of injury to the cervical spine are not fully understood, because the human cervical spine is an anatomically complex structure, subjected to a variety of loading conditions in an accident. A mathematical model of the cervical spine will be a valuable tool in the study and understanding of the mechanical behaviour of this complex biological structure.

This report gives the results of a literature survey conducted as the first part of a project concerning the development of a mathematical model of the human cervical spine. The model must describe, in detail, the biomechanical response of the human head and neck to various impact situations, and it must incorporate injury mechanisms.

The survey concerned the aspects relevant for mathematical modelling of the human cervical spine. Included is the literature relevant to the functional anatomy and biomechanics, injury mechanisms, and mathematical models of the cervical spine, as well as experimental results usable for model validation.

Contents

1 Introduction

The epidemiology of injuries reveals that injuries of the cervical spine are quite commonly found both in traffic accidents (mainly car crashes) and in leisure- and sports-activities. The cervical spine is particularly vulnerable in low-speed rear-end car collisions, which may result in so-called whiplash injuries.

The human cervical spine is an anatomically and mechanically complex structure that has much mobility. As it is a biological structure, large variations in geometry and mechanical behaviour of the cervical spine are found. Because of its complex nature and the variety of loading conditions encountered in accidents, the mechanisms of injuries to the cervical spine are not fully understood, even by specialists working in the field.

In the study and understanding of the biomechanical behaviour of the cervical spine, both mathematical modelling and experimental research are used. Experimental research on the (injury) biomechanics of the cervical spine has been conducted using human volunteers, intact human cadavers, head-neck and neck specimens, (motion) segments of the cervical spine, isolated spinal components, and animals. These experiments have been conducted to obtain insight into the functional anatomy and biomechanical behaviour of the normal and injured cervical spine, physical properties of the cervical spine and its components, the mechanisms of and tolerances to injury, and the responses of head and neck to impact situations.

Mathematical models can be used to simulate the behaviour of a biomechanical system in different (experimental) situations and can be used to obtain information that cannot be obtained from experiments. Mathematical modelling the mechanical behaviour of biological systems has proven to be a valuable tool in other fields of research as well as in the research on spine biomechanics.

Aim of Study

At the Eindhoven University of Technology, a research project is going on, concerning the development of a mathematical model of the human cervical spine. This model must describe, in detail, the biomechanical response of the human head and neck to various impact situations, and it must incorporate injury mechanisms.

According to Ward and Nagendra [115], the major pitfalls in mathematical modelling of biological systems are: oversophistication, lack of good physical properties data, and lack of validation. If a model includes (too) many details, the effect of these details on the behaviour of the model may be difficult to retrieve. During the process of modelling, usually numerous assumptions and simplifications have to be introduced, partly due to the lack of reliable physical properties data and partly to reduce complexity of the model.

To check on the assumptions used, a model has to be validated. Validation is achieved by correlating numerical predictions with experimental results covering the situations for which the model will be used.

Within this project, no experimental research will be carried out. Instead, results from experimental research, reported in the literature or obtainable elsewhere, will be used to develop a mathematical model of the cervical spine.

Scope of Survey

This report gives the results of a literature survey conducted as the first part of the project. The survey concerned the aspects relevant for mathematical modelling of the human cervical spine. Included is the literature relevant to the functional anatomy and biomechanics, injury mechanisms, and mathematical models of the cervical spine, as well as experimental results usable for model validation.

The functional anatomy and biomechanics of the cervical spine is the subject of Chapter 2. For the mathematical model, geometrical, inertial, and material characteristics of the biomechanically relevant components of the cervical spine are needed. These data together will be referred to as physical properties data. A review on studies on the physical properties of the cervical spine and its components is included in Chapter 2. Injury mechanisms and injuries (injury biomechanics) of the cervical spine are described in Chapter 3. A review of mathematical models of the (cervical) spine is provided in Chapter 4, which includes also a review on experimental data that can be used to validate a mathematical model.

2 Anatomy and Biomechanics

In this chapter, the functional anatomy and biomechanics of the mechanically relevant components of the cervical spine are briefly discussed. These components are the vertebrae, intervertebral discs, facet joints, ligaments and neck muscles. Sections 2.1 and 2.2 give a qualitative description of biomechanical behaviour of the cervical spine, whereas Section 2.3 deals with the quantitative aspects important in modelling the cervical spine. Additional and more detailed information can be found in References [48,53,58,59,86, 100,101,111,116], which were used throughout Sections 2.1 and 2.2, and those cited in Section 2.3.

2.1 The Cervical Spine

The human spine can be divided into four regions: the cervical, thoracic and lumbar spine and the sacroiliac region. The cervical spine is the upper part of the spine. It supports the head and protects the spinal cord. It is an articulate structure made up of joints, allowing for motion of the head relative to the torso. The basic four motions of head and neck are *flexion* (forward bending), *extension* (rearward bending), *rotation,* and *lateral bending* (sideward bending), see Figure 2.1.

The cervical spine comprises seven bony elements, called vertebrae (Fig. 2.2). The vertebrae are joined by soft tissues, anteriorly by ligaments and intervertebral discs, posteriorly by ligaments and facet joints. Facet joints and intervertebral discs carry load from one vertebra to another, whereas ligaments and neck muscles stabilize the cervical spine. Thus, vertebrae, intervertebral discs, facet joints, ligaments and muscles are the components that are biomechanically relevant, and they will be discussed in detail hereafter. Components

FLEXION · EXTENSION · LATERAL BENDING · ROTATION
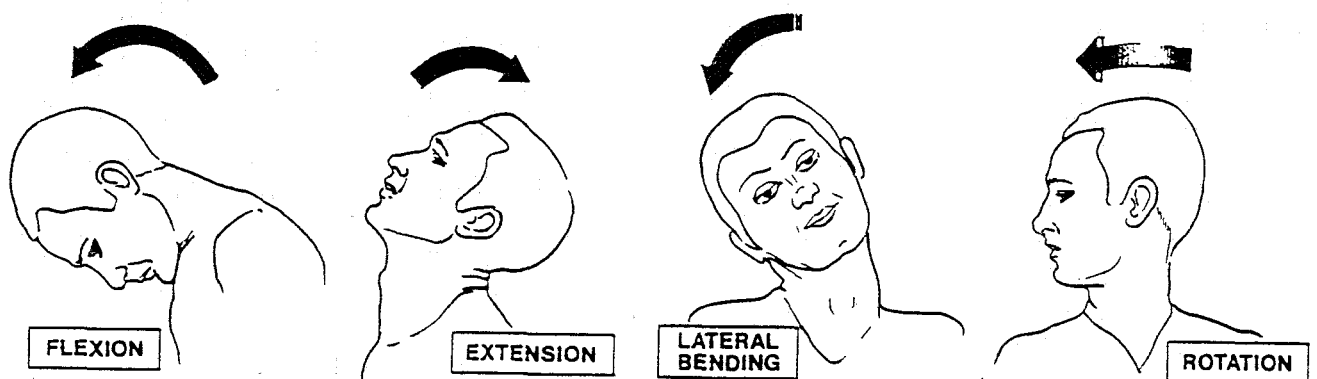
Figure 2.1: Basic motions of head and neck [48].
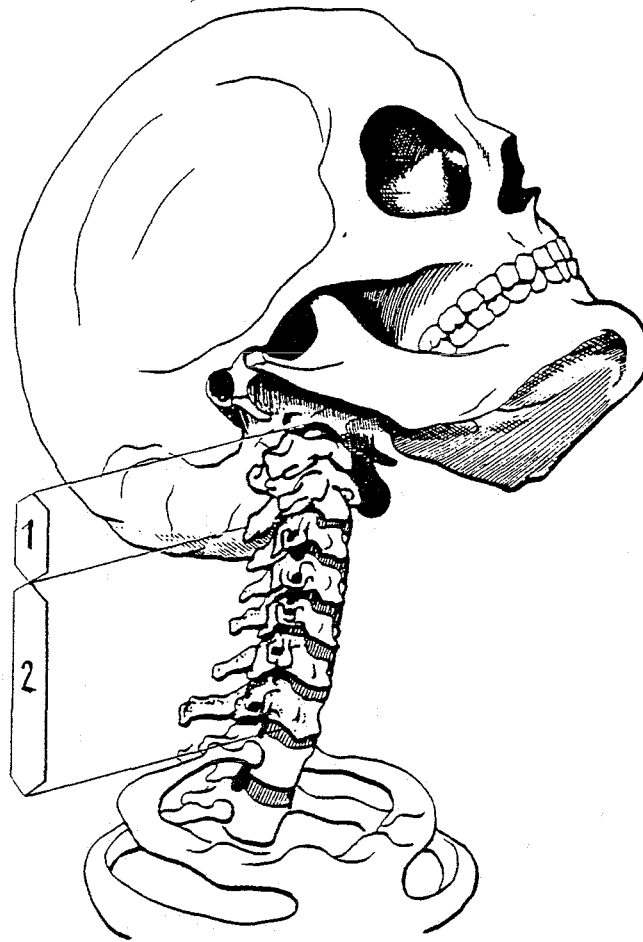
Figure 2.2: The upper (1) and lower (2) cervical spine, and skull [58].

like blood vessels, skin and fat are not taken into account, since these are of minor or no biomechanical importance. Despite their biomechanical irrelevance, the spinal cord and associated structures are included, since consequences of injury to the spinal cord range from minor disabilities to full permanent paralysis and even (immediate) death.

2.1.1 Cervical Vertebrae

The cervical spine comprises seven vertebrae, numbered C1 through C7. The lower five cervical vertebrae (C3-C7) are basically the same and are referred to as *typical vertebrae*. They gradually increase in size from C3 down to C7. The first (*atlas*) and the second cervical vertebra (*axis*) are distinct from each other and from the lower five. Due to these differences, the cervical spine can be divided into the lower cervical spine, comprising vertebrae C3 through C7, and the upper cervical spine, comprising axis, atlas and *occiput* (base of the skull, abbreviated as C0). The upper cervical spine is also called the occipito-atlanto-axial region.

Typical vertebra Generally, a vertebra consists anteriorly of a vertebral *body* and posteriorly of a vertebral *arch* (Fig. 2.3a). The body is a cylindrically shaped bone consisting of a centre of spongy bone surrounded by a thin layer of compact (cortical) bone. The *endplates* above and below the centre of the body are cartilaginous (hyaline cartilage).

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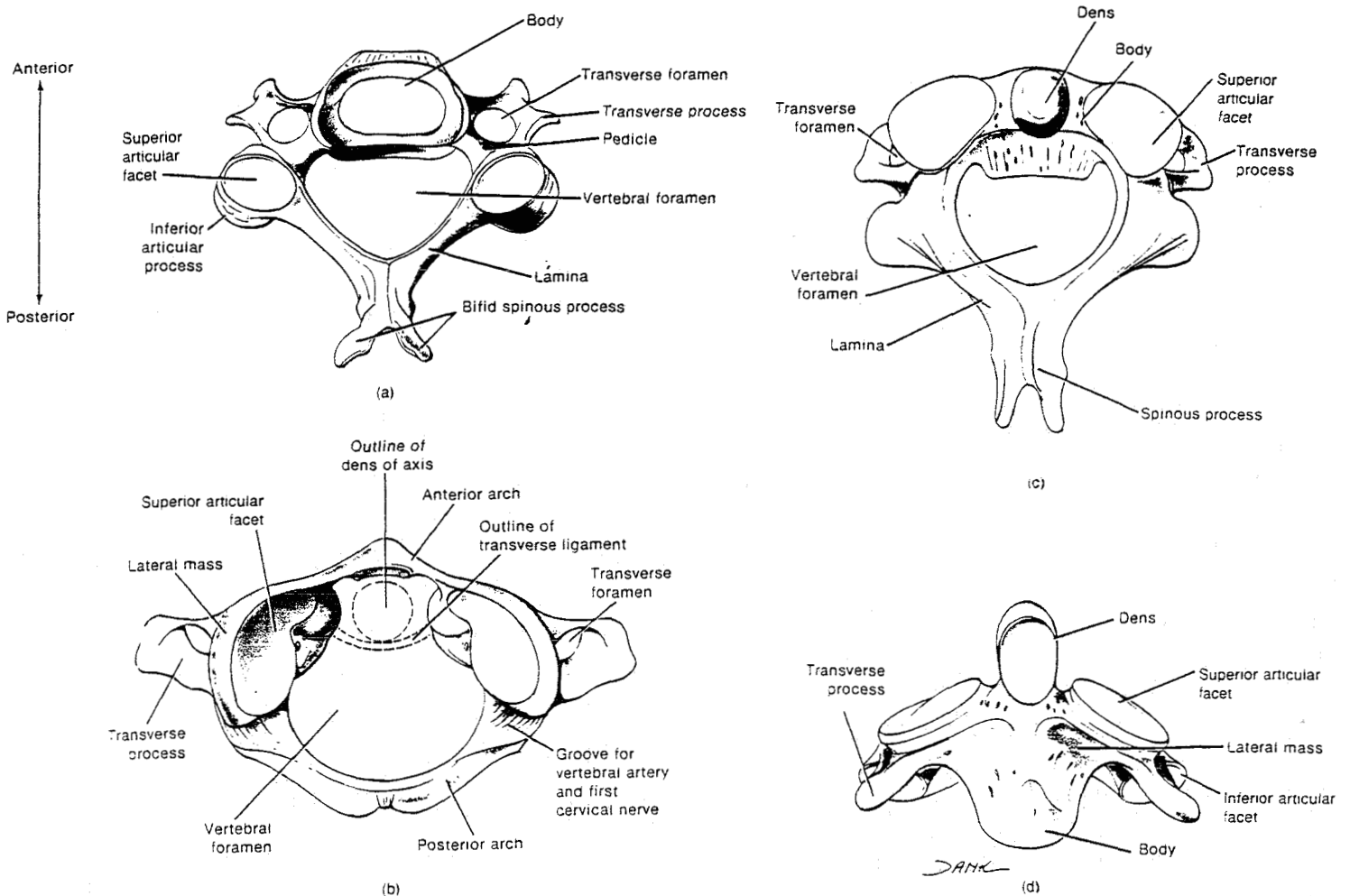
Figure 2.3: Cervical Vertebrae. (a) Superior view of a cervical vertebra. (b) Superior view of the atlas. (c) Superior view of the axis. (d) Anterior view of the axis [111].

The arch comprises two *pedicles*, two *laminae*, two pairs of *articular facets*, a *spinous process* and two *transverse processes*. The articular facets are found at the junction between the pedicles and laminae; they are part of the facet joints. The transverse and spinous processes act as attachment points for muscles and ligaments. Each transverse process contains a *transverse foramen* through which the vertebral artery and its accompanying vein and nerve fibres pass. The arch encloses the vertebral foramen. The vertebral foramina of all vertebrae together form the vertebral *spinal canal* through which the spinal cord and associated structures run.

Atlas and Axis The atlas is the first cervical vertebra (C1), and supports the skull. It has no vertebral body, but consists of a bony ring with anterior and posterior arches (Fig. 2.3b). The articular facets are large lateral masses situated on the arch, which comprises also the transverse processes. The axis is the second vertebra (C2). Like the lower vertebrae, it comprises a vertebral body and an arch, but it has an additional element, the odontoid process or the *dens* (Fig. 2.3c,d). The dens points out superiorly from the vertebral body of C2 and is, in fact, the missing body of the atlas fused to the

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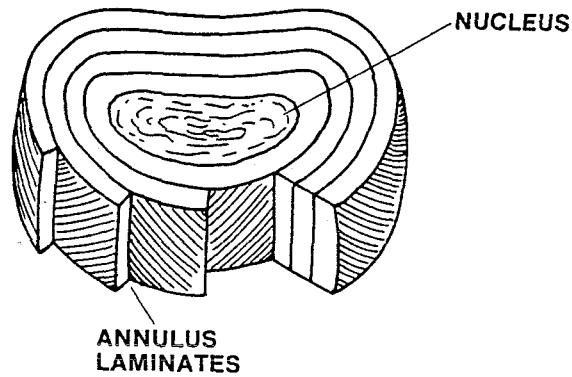
Figure 2.4: The intervertebral disc [116]

axis. The dens articulates with the anterior arch of the atlas through a synovial joint. Held firmly in position by ligaments, the dens makes a pivot around which atlas and head rotate, thus permitting axial rotation of the head.

2.1.2 Intervertebral Discs

There are two types of joints between two adjacent vertebrae: a fibrocartilaginous joint, known as intervertebral disc, and a synovial joint, known as facet joint. The intervertebral disc and a pair of facet joints form the load bearing junctions between two vertebrae.

The *intervertebral disc* is a fibrocartilaginous joint between the endplates of two adjacent vertebral bodies. It can be divided into two main regions, the *nucleus pulposus* and the *anulus fibrosus* (Fig. 2.4). The highly elastic fluid-like nucleus pulposus is located near the centre and is surrounded by a laminar set of spirally wound fibrous layers, the anulus fibrosus. The anulus fibres, which are firmly attached to the endplates of the vertebral body, run in a cross-like manner between any two adjacent layers.

The intervertebral disc is a strong joint allowing for motion between vertebrae in all directions. Discs support bending loads, absorb compression, and resist rotation, tension, and shear.

Discs are found between each of two adjacent vertebrae, except for occiput and atlas, and atlas and axis, between which no disc is found. In the cervical spine, the discs are relatively thick compared to the height of the bodies. At least *in situ* (which may be due to pretension of the cervical ligaments), discs are wedge-shaped: they are thicker anterior than posterior. As a consequence, the cervical spine is not straight, but curved convex anteriorly. This curve is known as *cervical lordosis*.

2.1.3 Facet Joints

Another load bearing junction between vertebrae is a pair of facet joints. Facet joints are formed by the articular facets of which each vertebra has four: two superior and two inferior to the arch. The facets articulate with corresponding facets of the vertebrae above and below.

The facets of the lower vertebrae are quite flat and symmetrical. The superior facets face upward and backward, the inferior facets forward and downward. Their inclination

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is about 45 degrees from front to back in the horizontal plane. The facets are enclosed by capsular ligaments to form facet joints, which are synovial joints. In general, synovial joints allow for sliding movements only, but within the facet joints more movement is possible due to the laxity of the articular capsules.

The concave and oval superior facets of the atlas articulate with the *occipital condyles*, the convex articular surfaces at the occiput. The geometry of this articulation is such that axial rotation of the head is not allowed: the two structures move as one unit in axial rotation. But it does permit the flexion/extension-movement seen when nodding the head. Atlas and axis articulate through the facet joints and the dens, which together allow extensive axial rotation.

2.1.4 Ligaments

Ligaments and muscles are the most important stabilizing components of the cervical spine. Ligaments provide stability for the bony elements and help preserve the articulate nature of the spine. Furthermore, they allow a requisite amount of spinal motion within physiologic limits, and prevent excessive motion to protect the spinal cord. Ligaments connect adjacent vertebrae or extend over several segments. Some ligaments run over the entire (cervical) spine, others are found between adjacent vertebrae of the lower cervical spine, and some ligaments are unique to the upper cervical spine.

Spinal ligaments are composed primarily of two substances: elastin (yellow elastic tissue) which allows ligaments to stretch and to return to original length, and collagen fibres for tensile strength. Ligaments are uniaxial structures: they resist tensile loads, but fold when subjected to compression.

Ligaments that run over the entire spine

Three spinal ligaments run along the entire length of the spine: the *anterior longitudinal*, the *posterior longitudinal* and the *supraspinous ligament* (Fig. 2.5). The anterior and posterior longitudinal ligaments line the anterior and posterior aspects of the vertebral bodies: they are strong collagenous ligaments attached to the bodies and the discs. The supraspinous ligament joins the tips of the spinous processes. In the cervical region, this ligament is found as the ligamentum nuchae. The anterior longitudinal respectively posterior longitudinal and nuchal ligaments limit extension respectively hyperflexion of the spine.

Ligaments between adjacent vertebrae of the lower cervical spine

The ligaments of the lower cervical spine include the *interspinous ligament*, the *ligamentum flavum*, the *capsular ligament* and the *intertransverse ligament* (Fig. 2.5). The interspinous ligament is a thin, tough membrane located between adjacent spinous processes. It blends together with the supraspinous ligament where they are adjacent. The ligamentum flavum is a strong elastic band that connects adjacent laminae. It is composed primarily of elastin which prevents folding of the ligament into the spinal canal during (hyper)extension. The capsular ligaments enclose the facet joints. The intertransverse ligaments connect adjacent transverse processes and limit lateral flexion. The other ligaments are all posterior ligaments and serve to limit flexion.

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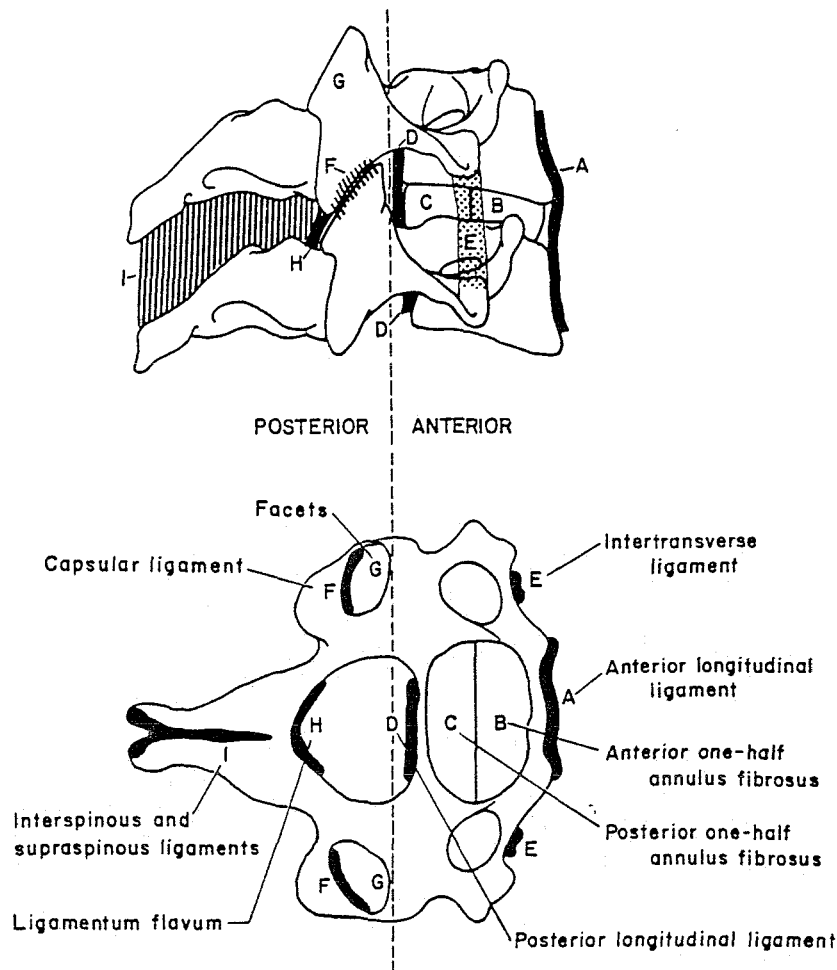
Figure 2.5: Ligaments of the spine and lower cervical spine [116].

Ligaments of the upper cervical spine

The ligamentous structures of the upper cervical spine include continuations of the lower cervical ligaments and a unique set of ligamentous structures well-suited of the functional requirements of this region. The most important ligaments are described below (Fig. 2.6).

The *tectorial membrane* is the upper extension of the posterior longitudinal ligament from the atlas to the base of the skull. Likewise, the *anterior atlanto-occipital membrane* is the continuation of the anterior longitudinal ligament and the *posterior atlanto-occipital membrane* is the continuation of the ligamentum flavum.

The *transverse atlantal ligament* is the transversal part of the cruciate ligament of the atlas. It is a strong horizontal ligament that holds the odontoid process against the anterior arch of the atlas: it constrains the dens posteriorly. The dens is hold anteriorly by the apical and alar ligaments. The *apical ligament* is a midline structure extending from the top of the dens to the occiput. The *alar ligaments* extend from the dens on each side of the apical ligament to the medial side of the occipital condyles. Apical and alar ligaments tend to limit rotation of the upper cervical spine.

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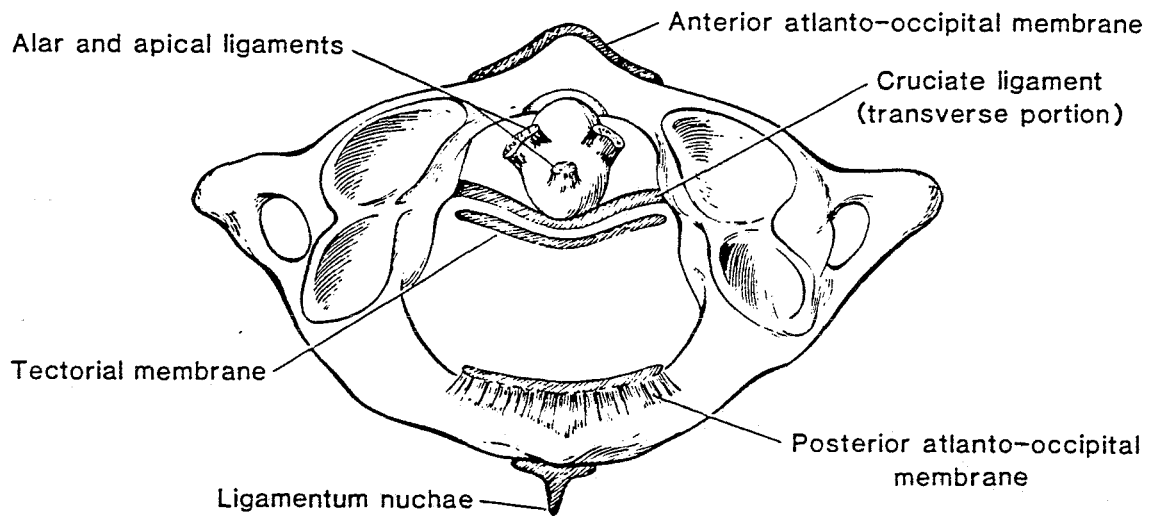
Figure 2.6: Ligaments of the upper cervical spine [100].

2.1.5 Neck Muscles

Neck muscles provide stability additional to the ligaments. Muscles are needed to maintain the head and neck in a given posture and to produce movements of the head and neck during physiologic activity. Although the muscles found in the cervical region are all under the direct control and will of the individual, they are also subjected to reflexive action.

The neck muscles can be divided into superficial, intermediate and deep muscles. Superficial neck muscles cross through the neck region without having a direct attachment to the cervical vertebrae. Intermediate neck muscles pass through the cervical region and have some attachments in the cervical region. Deep muscles join one vertebra to another or span one or more vertebrae. They act as stabilizers of the vertebrae and hold the head and neck in an upright position.

2.1.6 Spinal Cord

The central nervous system is formed by the brain and the spinal cord. The brain, lying within the skull, extends downwards to form the spinal cord, which lies within the vertebral spinal canal. This canal changes in length during physiologic flexion, extension and lateral bending. The changes in length are followed by the flexible, accordion like spinal cord: it unfolds in flexion and folds in extension. The cord is separated (protected) from the bony vertebral canal and its connecting ligaments by a fatty connective tissue and the meninges. Further, both brain and cord are surrounded by cerebrospinal fluid, which serves to protect these structures from traumatic forces. Spinal nerves arise from the spinal cord and make their exit between adjacent vertebrae.

2.1.7 Motion Segment

A motion segment comprises two adjacent vertebrae together with surrounding soft tissues: intervertebral disc, facet joints and associated spinal ligaments. It is the smallest unit exhibiting biomechanical features similar to the entire spinal column. Since the spine

Ranges of Rotation of the Cervical Joints			
	type of motion (representative angle in degrees)		
cervical joint	combined flexion/extension	one side lateral bending	one side axial rotation
C0-C1	25	5	5
C1-C2	20	5	40
C2-C3	10	10	3
C3-C4	15	11	7
C4-C5	20	11	7
C5-C6	20	8	7
C6-C7	17	7	6
C7-T1	9	4	2

Table 2.1: Representative values for the ranges of rotation of the cervical joints [116].

may be considered as a structure composed of motion segments connected in series, its total behaviour is a composite of individual motion segment behaviour. Therefore, motion segments are often used to study the biomechanics of the (lower cervical) spine. Due to its functional arrangement, the upper cervical spine is usually treated as a single biome-chanical unit and, similar to motion segments, subject of biomechanics studies. In the literature, a motion segment is also referred to as a functional spinal unit or intervertebral joint.

2.2 Head and Neck Motions

The cervical spine is an articulate structure made up of joints allowing for motion of the head. Motion between one vertebra and another is allowed (and limited) by facet joints, intervertebral discs, ligaments, and neck muscles: the intervertebral disc allows for motion in all directions; the facet joints determine the type and characteristic axis of motion (coupling); ligaments allow normal motion and prevent excessive motion of vertebrae; muscles are the primary source of force resulting in motion of head and vertebrae.

The basic movements are flexion, extension, lateral bending and axial rotation. Recently, White and Panjabi [116] reviewed studies on the ranges of motion of the cervical spine. Their results on "representative values" for the ranges of rotation of the cervical joints are reproduced in Table 2.1.

In the lower cervical spine, each of the basic movements occurs between each intervertebral joint. Although the disc allows for a small amount of movement between any two adjacent vertebral bodies, all the vertebral movements together account for the large mobility of the cervical spine as a whole. The flexion/extension movement is a combination of translation and rotation of one vertebra with respect to another. Due to the spatial orientation of the facet joints, lateral bending is associated with axial rotation such that when head and neck are bent to the right the spinous processes go to the left. This effect is known as a coupling: rotation (translation) of a body about (along) one axis is consistently associated with a simultaneous rotation about or translation along another axis [116].

Due to the unique shapes of the vertebrae and articulations, there is extensive rotation at the atlanto-axial joint. Roughly half of the axial rotation of the neck occurs at this joint and the remainder half occurs at the joints of the lower cervical spine. Axial rotation of the atlas is associated with vertical translation, due to the shape of the articulations (coupling). The occipito-atlantal joint allows for extensive flexion-extension movement.

2.3 Biomechanics and Physical Properties

The biomechanical behaviour of the lower cervical spine can be studied by using cervical motion segments. Similarly, the behaviour of the upper cervical spine can be studied by taking the occipito-atlanto-axial complex as a functional entity. Since the behaviour of motion segments is dependent upon the behaviour of its components, these components should be studied too. Therefore, physical properties of both motion segments and components are needed to study the biomechanics of the cervical spine quantitatively. Physical properties include the geometrical, the inertial and the material characteristics of cervical components and compound structures thereof. Regarding these properties, the cervical spine is assumed being symmetrical with respect to the sagittal plane, which is approximately the case.

2.3.1 Geometrical Characteristics

Geometrical characteristics include (1) the dimensions of vertebrae, articular facets, discs, ligaments, and muscles; (2) the locations of the places where the soft tissues are attached to the vertebrae; and (3) the compound configuration of all elements. Much of these information may be collected from detailed (cross-sectional) anatomy books, X-rays photographs, computed tomography scans (CT-scans), magnetic resonance imaging (MRI's) and skeletal material.

Further, studies dealing with these characteristics have appeared. Francis [34,35] studied (variations in) the dimensions of cervical vertebrae and their articular facets from human skeletal material of young adults. Francis measured: the transverse and anteroposterior length of the vertebrae; the transverse and anteroposterior diameter of the vertebral bodies, dens, vertebral foramen, and articular facets; and the height of the vertebral bodies and dens. Nissan and Gilad [85] describe an anthropometric model for human vertebrae in the mid-sagittal plane. The mid-sagittal appearance of vertebrae is idealized by a square-box approximation for the vertebral body to which a triangular shaped 'arch with spinous process' is attached. The shape of the articular surfaces is not considered at all. Anterior and posterior disc height is measured as the distance between two adjacent vertebral endplates of two adjacent vertebrae. Parameters were measured from lateral X-rays of the cervical (and lumbar) spine and averaged results (for over 130 healthy men) for vertebrae C2-C7 (and L1-L5) are given. Liu et al. [64] determined the geometry of cervical vertebrae of young males by measuring the coordinates of premarked points (36 per vertebra) relative to the vertebral body centre. Furthermore, the orientation of the articular facet joints and the articular facet-to-centre area ratios were obtained. The data on the measured coordinates are given in Reference [66].

Methods to reconstruct the three-dimensional geometry of (lumbar) vertebrae have been developed by Lavaste et al. [62] and Breau et al. [13]. Based on a morphological analysis, Lavaste et al. developed a method to describe the entire three-dimensional geometry of lumbar vertebrae. This method needs six geometrical parameters which can easily be obtained from lateral and frontal X-rays. With these parameter the other dimensions of the vertebrae are calculated to reconstruct its geometry. Breau et al. developed a method to reconstruct the three-dimensional geometry of (lumbar) vertebrae from CT-scans.

Goel et al. [39] obtained, for three cadavers, the origins and insertions of all the major muscles of the head-neck region with respect to both anatomical and global reference planes. They also measured weight, natural length and maximum width of each muscle.

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2.3.2 Inertial Characteristics

Inertial characteristics include mass, location of centre of gravity, and (principal) moments of inertia of the head, neck and vertebrae. The characteristics assigned to the vertebrae should represent those of a complete neck.

Data on the inertial properties of the head and the head and neck, have been reported by Beier *et al.* [6] and Walker *et al.* [114]. Liu *et al.* [67] determined the inertial properties of horizontal segments of a cadaver trunk. Since each segment contained one vertebra, the properties assigned to the vertebrae are those of neck segments at the level of the vertebrae. Liu *et al.* reported that large errors were encountered in the moments of inertia data for the cervical segments. So their results should be interpreted with care.

2.3.3 Material Characteristics

The *in vivo* characteristics of the material behaviour of cervical components are the most difficult characteristics to obtain. As the geometrical and inertial characteristics, they are subjected to intra- and inter-individual variation (with age, sex, race and physical condition). Moreover, the mechanical behaviour of biological tissues is nonlinear, non-homogenous, anisotropic and viscoelastic. Viscoelasticity means that the material exhibits creep, relaxation and hysteresis, and that the strength of the material increases with strain rate. Furthermore, material behaviour cannot be obtained from the living individual and the (*in vitro, in situ*) behaviour of cadaveric material differs from the *in vivo* situation. If the materials are not tested *in situ*, determination of the characteristics is further complicated by the difficulty in removing the structures without changing their material properties.

Mathematically, material properties are specified by constitutive equations. The unknown parameters of these equations have to be estimated from experimental results to obtain a valid model of specific material behaviour. Experimentally, material behaviour is presented by means of force-displacement (or stress-strain) curves, stiffnesses, load and deformation at failure, and so forth.

Force-displacement curves for biomechanical structures as motion segments or ligaments typically have the nonlinear, sigmoidal shape represented in Figure 2.7. The curve is divided into a physiologic and a traumatic range. The physiologic range starts with the neutral zone in which little force is needed to deform the structure. At the end of this zone, the stiffness increases substantially. Within this elastic zone, stiffness usually remains fairly linear. The traumatic zone starts when microtrauma occurs within the structure (like failure of ligament fibres). This start may be somewhat unclear for the particular structure under study. The traumatic zone ends when a (substantial) drop in force occurs: failure of the structure (like ligamentous disruption). Load and deformation at this point of failure are then defined as ultimate or failure strength of the structure.

Although stiffness is easily defined as the ratio of (incremental change in) force on and (related incremental change in) deformation of the structure, the nonlinearity of material behaviour gives rise to difficulties. In most publications, the experimentally obtained force-displacement curves are not reproduced in the paper, but represented by a stiffness coefficient. However, the load-displacement curves are nonlinear and thus difficult to describe by just a stiffness coefficient. Moreover, different stiffness calculations have been used by different authors. For example, for the curve given in Figure 2.7, stiffnesses have
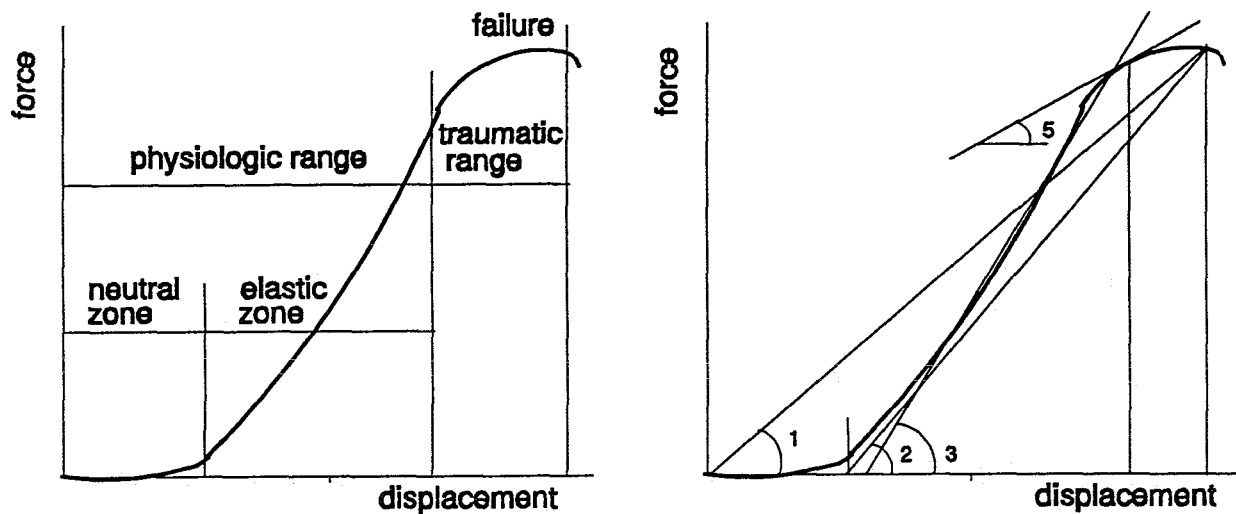
Figure 2.7: Typical force-displacement curve for biomechanical structures.

been calculated as: (1) the ratio of (maximum) applied force and deformation at this force; (2) the ratio of (maximum) applied force and deformation at this force minus the neutral zone deformation (3) slope of the most linear part of the curve; (4) stiffness calculated from linear regression analysis of the curve; or (5) slope of the curve at a certain load (or deformation). Stiffness calculated by (2) or (3) is usually named "average stiffness". Calculation (5), the exact definition of stiffness for a point of the curve, may be used to represent a complete curve if stiffnesses are given for a sufficient number of points. For all definitions, the load at which or the loading range for which the stiffnesses are calculated should be given.

Numerical models of and data on the mechanical behaviour of the cervical components are needed, if these components are modelled separately. In addition to the characteristics of individual tissue, the behaviour of cervical motion segments is of equal importance, since the (compound) behaviour of the components working together can not be studied from individual components alone. If not enough data can be collected from the literature, lumping of the mechanical behaviour of the soft tissues onto "an intervertebral joint" may be a valuable approach. Then, only experimentally obtained data describing the behaviour of motion segments is needed. In the following sections, the literature concerning the material characteristics of cervical spine segments and components is reviewed.

2.3.4 Material Characteristics of Cervical Spine Segments

Up to 1983, three-dimensional studies on the biomechanical properties of spine segments had been limited to the thoracic and lumbar regions [91]. To date, biomechanical properties of cervical spine segments have been examined by various investigators. In most studies, (motion) segments of the lower and upper cervical spine have been used to characterize experimentally the biomechanical behaviour of the cervical spine. In general, the experimental procedure is as follows (Fig. 2.8). The lower vertebra is fixed and loads are applied to the upper vertebra, while the resulting (main and coupled) displacements of this vertebra are measured. (The main displacement is the displacement in the same direction as the applied load, whereas the other displacements are referred to as coupled displacements.) Loads are applied statically (incrementally), quasistatically (with low

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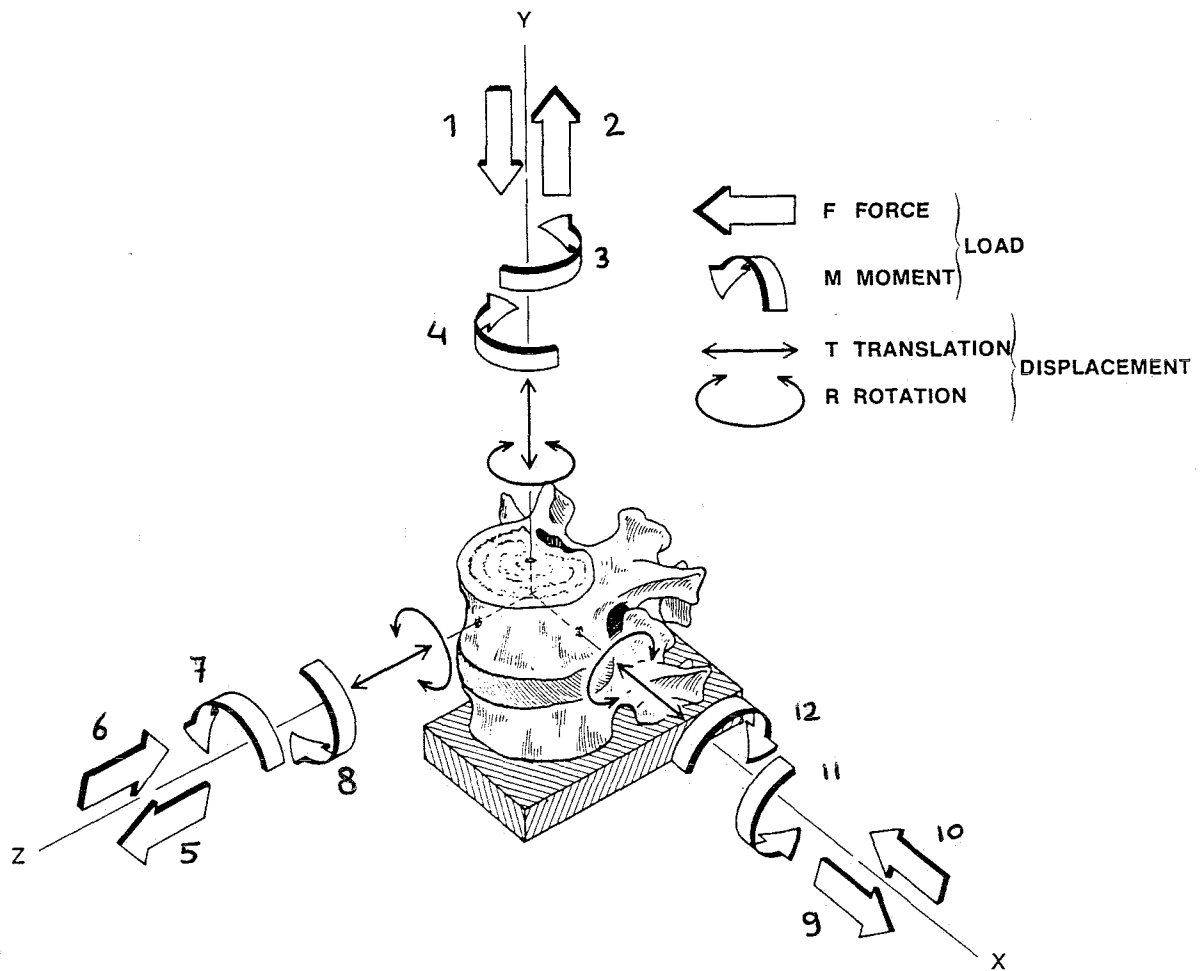
Figure 2.8: Motion segment with coordinate system to define the directions of load and displacement: (1) compression, (2) tension, (3) left and (4) right axial rotation, (5) anterior and (6) posterior shear, (7) left and (8) right lateral bending, (9) left and (10) right lateral shear, (11) flexion, (12) extension. [116]

deformation rate) or dynamically. Motion segment stiffness is then calculated from the measured force-displacement curves. Unfortunately, different authors have uses different stiffness calculations, which complicates a good comparison of reported stiffness values.

Panjabi et al. [91], Moroney et al. [81] and Shea et al. [105] have tested segments of the lower cervical spine to obtain biomechanical properties (load-displacement responses, strength or stiffness) and coupling characteristics of the lower cervical spine. The biomechanical properties of the upper cervical spine have been studied by Panjabi et al. [89] and Goel and co-workers [16,38,40]. White and Panjabi [116] collected average stiffness coefficients of a "representative" functional spinal unit of all regions of the spine for all modes of loading.

Lower Cervical Spine Studies

Panjabi et al. [91] subjected (18) functional spinal units to six different types of translational force: right and left shear, axial compression and tension, and anterior and posterior

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shear. Moroney *et al.* [81] tested (35) motion segments in compression, (anterior, posterior and lateral) shear, flexion, extension, lateral bending, and axial torsion. The latter used intact segments and disc segments which had their posterior elements (arch with ligaments) removed. They also measured the moments required for failure of the segments in flexion, extension or right lateral bending. In both studies, the lower vertebra is fixed, while the upper one is free to move in response to the loads applied to it. The load is applied statically: it is increased incrementally up to some maximum force (or moment) low enough to prevent injury (less than 80N or 2.2 Nm). The segment was given time to relax, before translational and rotational displacements of the upper vertebral body were measured. Thus, static characteristics of the segments are measured. Both Panjabi *et al.* and Moroney *et al.* did not find any systematic variation of the motion segment stiffness with vertebral level, therefore they calculated average stiffnesses for each type of loading. In both studies, large variations in mechanical properties of the segments were observed. Moroney *et al.* showed that these variations could not be reduced significantly by normalizing of the results with respect to disc dimensions. They also give an extensive discussion on the differences between their experimental method and results obtained and those of Panjabi *et al.*

Shea *et al.* [105] subjected spine segments to compression, tension, anterior and posterior shear, flexion and extension. They used (27) spine segments consisting of three adjacent vertebrae and their interconnecting ligaments and discs. The lower vertebra was forced to move, while displacements of the body of the middle vertebra were measured. Forces were measured at the upper vertebra, which was rigidly attached to a load cell. Segments were first tested non-destructively to obtain load-displacement data, and then loaded to failure in flexion or in flexion-compression. In some of the experiments, the upper vertebrae was given a fixed initial axial rotation. Quasistatic loading was applied with rates up to 5 mm/s translation or 5 deg/s rotation. Load-displacement hysteresis curves were generated to determine the specimen's response for complete cycles of loading. They found that the load-displacement curves were nonlinear for even small applied loads: stiffnesses are low near zero displacement and increase drastically before reaching failure. The loads applied were substantially higher (one order of magnitude) than those of Panjabi *et al.* and Moroney *et al.* and the calculated stiffnesses (defined as the slope of the curve at a specific relatively high load) were also higher.

Upper Cervical Spine Studies

Panjabi *et al.* [89] studied the three-dimensional physiologic motions of the C0-C1 and C1-C2 joints. Cervical spine specimens were loaded at the occiput for which six pure moments were used: flexion, extension, left and right lateral bending and left and right axial rotation. Loads were incrementally applied up to a maximum moment (1.5 Nm) large enough to produce physiologic motion and small enough to prevent injury. All six motion components of vertebrae and occiput were measured after each load step. They reported displacements at maximum load and the average flexibility coefficients for both joints.

Goel *et al.* [38] conducted static experiments on C0-C5 specimen to quantify the moment-rotation characteristics across the ligamentous occipito-atlanto-axial complex. Motion of vertebra C5 was fully restricted, while loads (pure moments) were incrementally applied at C0 up to maximum moment (0.3 Nm). Loads were applied in flexion, extension, left and right lateral bending and left and right axial rotation. The three-dimensional positions

of C0, C1, C2 and C3 at no-load, after each loading step and after removing the final load were measured. Relative rotation between C0-C1, C1-C2 and C2-C3 and averaged moment-rotation curves are reported. They found that the moment-rotation relationship is highly nonlinear. Furthermore, they noted that, in the absence of muscular actions, only small loads are required to produce relatively large angular rotations. They conclude that "this is supportive of the notion that ligaments across the occipito-atlanto-axial complex are lax and that the head is, therefore, held firmly to the neck principally by muscular actions".

Goel and co-workers [16, 40] quantified the quasistatic and the dynamic response of the occipito-atlanto-axial complex to axial rotation. They used C0-C1-C2 specimen of which C2-motion was fully restricted. Loading (axial torque) was applied at C0 until failure of the specimen. Specimens were subjected to loading rates of 4, 50, 100 and 400 deg/s respectively. The moment-rotation curves were highly nonlinear, showing less resistance in the initial phase, followed by a sharp increase in resistance in the final phase up to failure. Average torque-rotation curves for all loading rates are given. They found that the stiffness of the specimens increased at higher loading rates, and that the angular rotation at failure did not show any significant variation with loading rate.

Complete Cervical Spine Studies

Experiments have also been conducted using intact cervical spine specimen, sometimes including the head too, e.g. References [73, 84, 95, 94]. These experiments have been conducted to gain insight into the overall biomechanical behaviour and injury mechanisms of the cervical spine. To obtain information about the behaviour of spine segments, these type of experiments are less suited. They may be used, however, to validate the behaviour of a spine model of which the material characteristics have been obtained from segmental and component behaviour. The aspect of validation is discussed elsewhere, see Section 4.4.

2.3.5 Material Characteristics of Cervical Spine Components

Yamada [122] reported strength characteristics of numerous biological materials (organs and tissues) obtained from human and animal cadavers. With respect to the locomotor system, mechanical properties of bone (compact and spongy bone and vertebrae), cartilage, intervertebral discs, ligaments, muscles and tendons are given. Yamada's data on vertebral bodies, intervertebral discs and a few other tissues, relevant with respect to the spine, have been reproduced in Reference [101]. Fung [36] used principles of continuum mechanics to describe the mechanical behaviour of bio-fluids and bio-solids (hard and soft tissues). Emphasis is put on constitutive equations that can be used to describe the behaviour of biological tissues. Constitutive equations for, among others, muscles, bone and cartilage are discussed.

Vertebrae, intervertebral discs and facet joints

Yamada reported ultimate strength and deformation data for (wet) cervical vertebrae and discs subjected to compression or tension. Stiffness data for cervical vertebrae have not been reported to date. Since the deformation of the vertebrae is small compared to the deformation of discs, vertebrae may be treated as rigid bodies. Vertebral deformation may be taken into account by transferring it to the disc characteristics.

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Disc stiffnesses for various types of loading have been reported by Moroney et al. [81], who used intact and disc segments, as is discussed above (Section 2.3.4). Their data on disc segments represent disc-ligament behaviour; for compression this is equal to disc behaviour. Pintar et al. [92] reported the biomechanical properties of cervical (and thoracolumbar) discs in tension.

To date, biomechanical properties of cervical facet joints have not been reported in the literature. However, the capsular ligaments of the facet joint have been studied, as is described below.

Ligaments

Force-displacement curves for spinal ligaments typically have the nonlinear, sigmoidal shape, represented in Figure 2.7.

Average failure strengths (load and deformation at point of failure) of the most important upper and lower cervical spine ligaments have been collected by White and Panjabi [116]. Chazal et al. [17] studied the geometrical and biomechanical properties of various spinal ligaments, subjected to quasistatic loading (1 mm/min). With respect to the cervical spine, data were obtained for the anterior and posterior longitudinal ligaments. Included are mean values for stress and strain at three characteristic points of the sigmoidal force-displacement curve. Dvorak et al. [25] reported the failure strength and the dimensions of the alar and transverse ligaments of the upper cervical spine. The ligaments were loaded quasistatically at a rate of 1.5 mm/s.

Myklebust et al. [83,93] reported average values for the failure strength of spinal ligaments of all spinal levels. Ligaments were tested in situ by sectioning all element except the one under study. Load was applied quasistatically at a rate of 1 cm/s until failure of the ligament occurred. Force-displacement curves typically exhibited a sigmoidal shape. With respect to the lower cervical spine, data for the anterior and posterior longitudinal and the interspinous ligaments, the ligamentum flavum and the joint capsules are reported. With respect to the upper cervical spine, data for the anterior and posterior atlanto-occipital membrane, the apical and alar ligaments, the vertical part of the cruciate ligament and the joint capsules are given.

Yoganandan et al. [123] investigated the in situ dynamic response of the anterior longitudinal ligament and the ligamentum flavum of the cervical spine. Four different loading rates (9, 25, 250, and 2500 mm/s) were applied to obtain the (nonlinear) displacement-force curves up to failure from which several biomechanical properties were determined. They reported results for force and displacement, and energy-absorbing capacity of the structure at failure, as well as for the stiffness of the ligaments, which was defined as the slope of the most linear part of the response.

2.4 Discussion

Geometrical and inertial characteristics are sufficiently available from the literature. Data on material characteristics of individual cervical spine components and of cervical motion segments are however limited and incomplete.

Stiffness for cervical vertebrae as well as the (compressive) stiffness of facet joints have not been reported, to date. Disc stiffnesses have only been reported for compression

and tension. The most important ligaments of the cervical spine have been studied under static and quasistatic loading. Dynamic characteristics have been obtained for the anterior longitudinal ligament and ligamentum flavum only.

Experiments have been conducted to obtain material characteristics for both upper and lower cervical spine segments. With respect to the lower cervical spine, characteristics were obtained for static loading (flexion, extension, compression, tension, axial rotation, lateral bending and anterior, posterior and lateral shear) and for quasistatic loading (flexion, extension, compression tension, and anterior and posterior shear). No dynamic experiments from which data on material characteristics may be obtained have been reported to date. Only in a few of these studies, loads were applied up to failure of the specimens. The experimental methods used and (therefore ?) the results obtained by various authors differ. With respect to the upper cervical spine, data from static experiments (with low maximum load) have been reported for flexion, extension, lateral bending and axial rotation. Results of quasistatic and dynamic applied loads (up to failure) have been reported for axial rotation only.

Data are obtained from cadaveric material, the properties of which differ from those of the living human. Furthermore, cadaveric material is often obtained from elderly persons and the material characteristics may not be representative for the "average" behaviour of the biological tissues (if such behaviour exists at all). Differences in mechanical behaviour due to differences in sex, race or physical condition are usually not considered at all.

The limited availability of and the observed variations in the physical properties of cervical spine segments and components, imposes difficulties and limitations on the development and subsequent validation of a cervical spine model which includes all anatomical components separately. To reduce the complexity of the model and the number of (unknown) material parameters in the model, component behaviour can be lumped together, representing motion segment behaviour. Then, only experimental data for motion segment behaviour is needed.

3 Injury Mechanisms and Injuries

The subject of this chapter is injury of the cervical spine. Cervical spinal injuries may occur due to direct impact to or inertial loading of the head and neck. In both cases, head and neck are forced to move forward, backward, sideways or to rotate relative to the torso, which may result in injuries to the head and cervical spine. Each of these loading combinations may represent a different injury mechanism and each can result in different injuries of the cervical spine.

Injuries and injury mechanisms of the cervical spine as well as a classification of injuries which relates a particular mechanism to an injury or group of injuries will be discussed. The literature on injuries and injury mechanisms of the (cervical) spine is extensive. Valuable retrospective studies include References [2,19,43,49,74,76,59,100,101,116], which were used throughout this chapter.

3.1 Epidemiology

The incidence of neck injuries reported in different epidemiological studies show wide variations. This is partly due to the criterion used to select cases from larger data collections. Major difference is the in- or exclusion of the non-injured persons involved in (car) accidents. With respect to the cervical spine, another difference is due to the group of minor injuries. These injuries are diagnostically difficult to define, since no identifiable structural injury of the cervical spine is found, despite the fact that the victim may suffer from a variety of serious but "vague" complaints, like headaches, neck pain, dizziness and forgetfulness. In some studies, victims suffering from the complaints mentioned are more easily classified as injured persons than in other studies.

If only car accidents are taken into account, the following incidence for cervical spine injuries have been reported. Including both injured and non-injured car occupants, Anderson et al. [3] found 2.4 % neck injuries of which 88 % were rated as minor injuries (AIS 1)[1], while Faverjon et al. [33] found 10 % neck injuries for front seat occupants only (90 % classified as AIS 1). Including only injured car occupants, Bunketorp et al. [15] reported an incidence of 25 % (94 % AIS 1), and Jorgensen et al. [54] one of 14.3 % (90 % AIS 1). Otte and Rether [88] found 1.3 % AIS 1 and 0.6 % AIS > 1 neck injuries. Vallet and Ramet [112] reported for all neck injuries a rate of 7 % and for neck injuries with AIS > 2 a (remarkably high) rate of 6.5 %. According to Lövsund et al. [68], the most common injury in rear-end car collisions is neck injury: 10 % of all occupants sustain neck injuries (of AIS 1).

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[1]The Abbreviated Injury Scale, a standardized system for injury severity rating, ranges injuries according to the "threat to life" on a scale from minor (AIS 1), via moderate (2), serious (3), severe (4) and critical (5) up to maximum (or lethal) injury (6) [5].

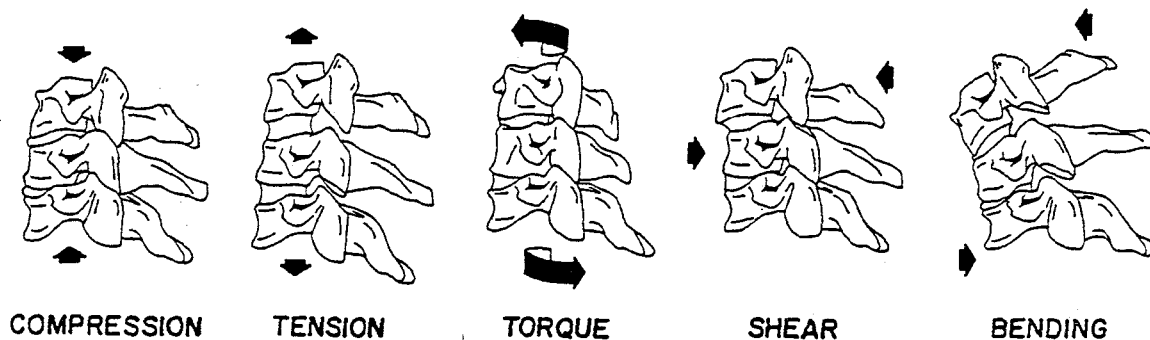COMPRESSION   TENSION   TORQUE   SHEAR   BENDING

Figure 3.1: Principal applied loading of motion segment.

Some general observations include (1) severe neck injuries are uncommon in car accidents, (2) the neck is more rarely and less severely injured than other body regions, and (3) neck injuries are mostly accompanied by injuries of other regions of the body (like thorax and head).

3.2 Injury Mechanism and Classification of Injuries

Injuries to the spine may result from a complex series of forces and moments applied to the body in a variety of different ways. The loads on the body are ultimately transmitted to the region of the spine where the injury takes place [116]. Cervical spine injuries are associated with external loads transmitted to the human head and neck. To describe the mechanism through which the injury took place, insight into the forces and displacements necessary to produce the injury is needed.

An injury mechanism should be based on local movements of and local forces on vertebrae and motion segments of the cervical spine, since injury is a localized event. The most important complex of forces is the one that actually causes the injury. It has been referred to as the major injuring vector (White and Panjabi [116]) or the principal applied load (McElhaney and Myers [74]) and it can be used to classify injury mechanisms. Then, injuries can be classified according to its injury mechanism.

Here, the classification of McElhaney and Myers, which is based on the principal applied loading of motion segments of the cervical spine, is used. They differentiate between five types of loading: compression, tension, torsion, shear and bending loads (Fig. 3.1). Their classification is given in Table 3.1. It is obvious from this scheme that groups of injuries may result from identical or similar injury mechanisms. Harris et al. [43] state "that it is reasonable to assume a direct relationship between the magnitude of causative force and the type of injury". Allen et al. [2] used this to range injuries from trivial to severe within a given injury mechanism. Another interesting aspect that can be seen from Table 3.1, is that injury mechanisms can be described essentially as one-dimensional (compression, tension, torsion) or two-dimensional (shear, bending) mechanisms. In reality, the injury mechanism is much more complex, due to other (minor) forces modifying the principal applied load and making the injury mechanism three-dimensional.

In the literature, injury mechanisms have also been related to the motion of the head. However, loads on and motions of the head do not always reflect the segmental conditions and patterns of injury at the level of the individual vertebrae [19]. Therefore, the following

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Injury Mechanisms and Injuries of the Cervical Spine	
principle applied loading	possible type(s) of injury
(vertical) compression	multi-part atlas fracture Jefferson fracture vertebral body compression fracture burst fracture
compression-flexion	vertebral body wedge compression fracture hyperflexion sprain unilateral facet dislocation bilateral facet dislocation
compression-extension	posterior element fractures
tension	occipito-atlantal dislocation
tension-extension	whiplash anterior longitudinal ligament tears disk rupture horizontal vertebral body fracture hangman's fracture
tension-flexion	bilateral facet dislocation
torsion	rotary atlanto-axial dislocation
horizontal shear	odontoid fracture transverse ligament rupture
lateral bending and shear	nerve root avulsion transverse process fracture
various	teardrop fracture

Table 3.1: Classification of injury mechanisms and injuries of the cervical spine based on the principal applied loading (Adapted from: [74]).

three points should be kept in mind, when relating (observed) motion of or loads on the head and neck to motion of or load on a motion segment, that is to a particular injury mechanism [74]:

1. the observed motion of the head relative to the torso may not be the same as regional motions in the cervical spine;

2. the observed head motion may have occurred after the injury took place (post-injury kinematics), and thus not reflect the true injury mechanism but motions of an unstable spine;

3. motion segment loads are not only related to the load applied to the head but also to the spinal configuration at the moment of impact.

Two-column concept of the spine

The two-column concept of the (cervical) spine is a valuable concept in understanding injury mechanisms. In this concept the spinal column is divided into an anterior and a posterior column. The anterior column includes the anterior and posterior longitudinal ligament, the vertebral body and the vertebral disc. The posterior column consists of all the spinal elements posterior to the posterior longitudinal element, including the facet joints.

The basic principle of this concept is that during flexion the anterior column is compressed whereas the posterior column is distracted. Thus, a flexion bending moment results in

compression of the intervertebral disc and vertebral body and in tension of the posterior elements, whereas an extension bending moment results in the opposite: tension of disc and body and compression of the posterior elements. Flexion or extension of the spine *additional* to compression (tension) of the spine will change the load pattern of vertebral segments and may therefore result in other injuries than found in pure vertical compression (tension) alone. For example, flexion in addition to compression of the spine will increase the amount of compression of the body and disc, but decrease the amount of tension in the posterior elements; and consequently, failure of anterior elements becomes more likely than failure of posterior elements.

3.3 Cervical Spine Injuries

Four types of injury can be distinguished: soft tissue injuries, dislocations, fractures and fracture-dislocation. Soft tissue injuries include disruption of ligaments and muscles and failure of intervertebral discs. Dislocations are changes in the normal anatomic relationship between bony structures; they occur secondary to ligamentous disruption. Fractures refer to any alteration in the bony structure of the vertebrae. Fracture-dislocation is a mixture of the latter two: it involves bony fracture and displacement of one or more vertebrae or bony elements from their normal anatomical locations.

The relative incidence of injuries is reported by Allen *et al.* [2] who studied 165 cases of lower cervical spine injuries. They divided the injuries among six injury mechanisms and reported the following relative incidences (in parentheses): tension-flexion (37%), compression-extension (24%), compression-flexion (22%), vertical compression (9%), tension-extension (6%) and lateral flexion (3%).

The following description of injuries is primarily based on the work of McElhaney and Myers [74], except where noted. The injuries are discussed per injury mechanism (based on segmental loads; Table 3.1) and the segmental loading is related (whenever possible) to commonly seen loading or motion of the head and neck. Thus, when reading this section, the earlier mentioned three points should be taken into account. Spinal cord injuries are discussed separately at the end of this section.

Vertical Compression

Compression may occur in a wide variety of accidents, like in diving accidents, American football accidents and car crashes. Pure vertical compression is infrequently encountered, because at the moment the force is delivered the cervical spine must be straight, that is neither flexed nor extended. The forces on the head are transmitted to the spine through the occipital condyles and they may result in upper and lower cervical injuries.

In the upper cervical spine, compression may result in a *multi-part fracture* of the ring of the atlas: a fracture of the ring into two or more parts. A multi-part fracture is referred to as a *Jefferson fracture* if there are two fractures (bilateral) in the anterior part of the ring and two in the posterior part.

In the lower cervical spine, compression may result in fractures of the vertebral bodies. A *burst fracture* is a multi-part fracture of the vertebral body. The other *compression fractures* include destruction of the cancellous bony centre with loss of disk height and fracture of the vertebral endplate with vertical herniation of the nucleus pulposus into the centrum.

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Compression-Flexion

Flexion occurs when the head and neck are forced forward on the torso. Compression-flexion of the neck occurs in accidents where the head and neck have to stop the moving torso and are, simultaneously, forced into flexion by an obstacle. This may happen, for example, when hitting the bottom of a swimming pool in diving. Compression-flexion injuries include vertebral body wedge compression fracture, unilateral and bilateral facet dislocation and hyperflexion sprain.

The *wedge compression fracture* is a failure in the anterior part of the vertebral body. It results from excessive compressive stresses due to a combination of a flexion bending moment and compressive loading of a motion segment.

Bilateral facet dislocation results from an anterior displacement of the superior vertebral body over its subjacent vertebra. All the ligamentous structures at the level of the injury are disrupted. This displacement results in a dislocation of both facet joints with subsequent locking of the facet surfaces in a tooth to tooth fashion. The inferior facets of the dislocated vertebra lie anterior to the superior facets of the subjacent segment. In *unilateral facet dislocation*, the displacement results in a dislocation and locking of only one facet joint. The mechanism of unilateral dislocation is similar to that of bilateral dislocation, with the difference being some form of asymmetric loading or deformation.

In *hyperflexion sprain* are the posterior ligamentous structures torn or stretched without producing dislocations or fractures of the facet joints. The mechanism is similar to facet dislocation, but with less extensive disruption of the motion segment. Hyperflexion sprain is an integral part of facet dislocation.

Compression-Extension

Extension may produce both upper and lower cervical spine injuries. The type of injury produced by extension bending moments (anterior ligamentous injuries or posterior bony injuries) depends largely on the type of loading of the head which produced the bending moment. Compression-extension, tension-extension and horizontal shear type of injuries may be produced.

Compression-extension may result in *posterior element fractures*. These fractures appear to be the result of direct bony impingement of the posterior elements against each other. Injuries include laminar fractures, pedical fractures, crushing injuries of the pars interarticularis, fractures of the spinous process or fracture of the posterior arch of the atlas [74,116]. The type of injury depends on the relative magnitude of the compression force and extension bending moment.

Tension

Pure tensile injuries appear to be restricted to the upper cervical spine. Excessive tensile loading (of the head) results in occipito-atlantal distraction with unilateral or bilateral dislocation of the occipital condyles. This *occipito-atlantal dislocation* tends to produce ligamentous injury without bony fracture.

Tension-Flexion

Tension-flexion is seen in frontal car crashes with belted occupants, where the torso is restrained and where the head and neck are free to move. During the accident, head and neck are forced to move, due to inertial loading. This results in a flexion moment that tends to separate vertebral segments and hence in tension-flexion loading of the neck.

Bilateral facet dislocation is also found as a result of tension-flexion loading. Thus, a flexion bending moment seems to be the primary base for this injury. Interestingly, Crowell *et al.* [19] classify unilateral and bilateral facet dislocation and hyperflexion sprain only as tension-flexion injuries.

Tension-Extension

Tension-extension injuries include whiplash, structural injuries to the anterior column and hangman's fracture. The *hangman's fracture* is a fracture of the pedicles or pars interarticularis of the posterior arch of the second cervical vertebra. The fracture separates the anterior from the posterior elements of the vertebra. It was typically found as a result of (judicial) hanging and is still found as a result of blows to the face and chin.

Whiplash is considered a hyperextension injury because of the extension bending moment resulting from inertial loading of the neck by the head. It typically occurs as a result of (moderate or severe) rear end collisions in automobiles. The injury is not associated with identifiable structural injury. Chronic symptoms of the injury include head and neck pain, muscle stiffness and tension, disequilibrium and emotional disturbances. Structural injuries to the anterior column include *anterior longitudinal ligament tears, disk rupture* and *horizontal vertebral body fracture*. These injuries may be produced in addition to whiplash at larger impact severities.

Torsion

Excessive torsional loading of the head may produce *rotary atlanto-axial dislocation*. This is a dislocation of one or both of the atlas surfaces on the axis facet joint surface, with or without tearing of the alar ligaments. Torsion injuries are restricted to the upper cervical spine, because the lower cervical spine is stronger in torsion than the atlanto-axial joint.

Horizontal Shear

Horizontal shear may be produced as part of a loading complex which forces the head and neck into flexion or extension. This shear can either produce *fracture of the odontoid process* or *rupture of the transverse ligament*. Dens fractures are more common than transverse ligament ruptures. The dens can be fractured at the tip, through its body or through the base in the body of the atlas.

Lateral Bending and Shear

Lateral bending rarely occurs as the dominant injury producing force to the cervical spine. It is more commonly seen as modifying the injury producing force [43]. Lateral bending and shear in combination with other loads are thought to produce the injuries already mentioned with a greater degree of trauma on the side to which the lateral loads are

directed. Furthermore, lateral shear and shear can produce *nerve root avulsion* and *transverse process fracture*.

Various Loadings

A *teardrop fracture* may result from various types of loading. It is a bony avulsion of the anteroinferior or anterosuperior portion of the vertebral body in the lower cervical spine. Teardrop fractures can occur in isolation or in association with ligamentous injury, vertebral burst fracture and intervertebral disk rupture.

Spinal Cord Injuries

Injuries can also be classified as clinical stable and instable injuries. Stable injuries are those in which there is no deformation, displacement, or fracture of vertebral body or disc, that may cause spinal cord damage or injury to nerve roots [59]. A clinical instable spine, on the other hand, has no longer the ability to maintain relationships between vertebrae in such a way that injury to spinal cord and nerve roots can be prevented, when the spine is subjected to physiological loads [116].

In principle, all of the injuries mentioned, may be associated with more or less severe spinal cord trauma, depending on the magnitudes of forces. As a consequence, all injuries may be classified as clinical instable injuries. However, only a few of the injuries are frequently associated with spinal cord trauma. Multiple-part atlas fractures are commonly associated with fatalities and disabilities. Lower cervical compression injuries may be associated with cord injury. Bilateral and, to a lesser extent, unilateral facet dislocations are frequently instable due to the narrowing of the spinal canal. Occipito-atlantal dislocation is usually lethal due to the distraction and subsequent transection of the spinal cord. Transverse ligament rupture and dens fracture are frequently instable and may result in (lethal) spinal cord impingement. Lateral shear and bending may result in nerve root avulsion injuries.

3.4 Injury Criteria

Injury may result from deformation of the cervical spine beyond normal (physiologic) ranges of motion and/or from excessive forces. These forces usually originate from inertia forces and cause the viscoelastic structures to deform rapidly (high strain rates) resulting in traumatic forces even at moderate deformation of the spine. Injury is thus related to both deformation and acceleration.

To define neck injury criteria, tolerance levels have to be determined. This is an extremely difficult problem, because of the wide variety of anatomical structures, injuries and loading conditions involved in an accident. Furthermore, muscular action may put loads into the system and influence both load and injury. The role of muscles with respect to injuries has not been studied in detail; it requires experimental research done with (anaesthetized) animals, since in (human) cadavers no muscular action is present.

To obtain tolerance limits, experimental biomechanical studies have been performed, using intact cadavers, (head-) cervical spine specimens, cervical motion segments and isolated components. Subjected to well controlled loading (or deformation), deformation of and load on the structure were measured and correlated with observed injuries. These studies have been reviewed recently by various authors, e.g. References [49,74,100,116]. Detailed

information may be obtained from these reviews and the references cited therein. A general observation is that the results from these experiments show a wide range of tolerance limits for all types of loading.

4 Mathematical Models

Mathematical modelling of biological systems has proven to be a valuable tool for studying the mechanical behaviour of such systems. Reviews on mathematical biomechanical models are provided by various authors. King and Chou [61] reviewed (in 1976) mathematical models related to the biomechanics of impact, whereas King [60] reviewed (in 1984) mathematical models of nonimpact type. Ward and Nagendra [115] described the (in 1985) current generation of biodynamic models. An extensive review on mathematical models of the spine and spinal components (up to 1986) was given by Yoganandan *et al.* [124]. The use of finite element analysis in orthopaedic biomechanics has been reviewed (in 1983) by Huiskes and Chao [50]. Mackerle [70] presented an extensive bibliography of recent publications (up to 1991) on the use of the finite element method in biomechanics.

In this chapter, mathematical models of the human cervical spine are reviewed. The four types of mathematical models of the human (cervical) spine that are distinguished, are pivot models, continuum mechanics models, discrete parameter models, and finite element models. Pivot models are reviewed in Section 4.2, the other models in Section 4.3. Section 4.4 deals with the aspects of validation.

4.1 Scope of Survey

Injury analysis requires that the mechanical behaviour of the cervical spine is represented in detail. In other words, the model must describe not only the *global* kinematics and dynamics of the head and neck, but also the *local* kinematics and dynamics of individual vertebrae and other relevant cervical components. The global motion of the head relative to the torso can already be described by relatively simple three-segment two-joint models. These two-pivot models have been developed primarily for describing this global head motion and they can be used to compare detailed models with. Therefore, pivot models are included in this review (Section 4.2). Pivot models cannot provide information about the local kinematics and dynamics of the neck. This information may be provided, however, by (cervical) spine models that include a detailed representation of the mechanical behaviour of the various anatomical structures of the human neck. These models should be capable of describing local deformations and forces (stresses and strains) that can be related to injury mechanisms and injury tolerance levels.

Thus, injury analysis requires detailed models. In the literature, three types of models that allow for a more detailed representation of cervical spine mechanics are found: continuum mechanics, discrete parameter and finite element models. First, a general description of these types of models is given in Section 4.3. Then, models of each type are reviewed. Emphasis is put on the more recent contributions on three-dimensional models in which the cervical spine is represented in detail. Finally, the two most advanced three-dimensional

dynamic models are discussed in detail.

Once a model has been developed, it has to be validated through comparison of predicted model response with similar results obtained from experiments. Both global and local behaviour have to be validated. Experiments that can be used for global or local validation of the model are reviewed in Section 4.4.

4.2 Pivot Models

4.2.1 Two-Pivot Models

The global head motion (motion of the head relative to the torso) can already be described by relatively simple three-segment, two-joint models. In these *two-pivot models*, the neck is modelled as a rigid or extensible link that connects the movement of the torso (at T1) to the head. Head and torso motion is determined from experiments (sled tests) with volunteers or cadavers. The experimental obtained torso motion is used in the model to predict head motion (angular and linear displacement, velocity and acceleration). This head motion is then compared with the experimental head motion to validate the model. Pivot models have been developed by various authors, amongst others, by Bosio and Bowman [12], Tien and Huston [110] and Wismans *et al.* [120,121]. Their modelling approaches will be discussed briefly.

Wismans *et al.* [121,120] used a two-pivot model to analyze the global head-neck kinematics and dynamics of human volunteers and cadavers subjected to sled acceleration tests. The model comprises three rigid links, representing head, neck and torso, connected by two pivots. Both pivots allow for rotation in the plane of impact, and the upper pivot allows also for rotation of the head around the neck axis. The upper pivot is located at the occipital condyles. The position of the lower pivot and the neck length were determined from approximation of the occipital condyles trajectories with a circular arc. The lower pivot location was found to be near the origin of the T1 coordinate system. Head-neck dynamics were analyzed to obtain torque-rotation characteristics for the pivots. These characteristics were all approximated by linear torsional stiffnesses. According to Wismans *et al.* averaging of the characteristics may be appropriate to obtain a general model, since both the effect of test severity on the characteristics and the variation of the characteristics between different subjects were small. No simulations performed with the model have been reported, however.

Tien and Huston [110] used a model comprising three rigid bodies, representing head, neck and torso, that are connected by springs and dampers, representing the soft tissue complex at these levels. The values for the parameters of the springs and dampers were found by curve fitting the numerical results onto experimental results of volunteer sled tests. In this way, Tien and Huston obtained a good match between the numerical and experimental found linear and angular head acceleration and velocity.

Bosio and Bowman [12] developed two-pivot models in which the neck is modelled as an extensible link, with elongation and compression properties. The neck has a pivot at the C7-T1 articulation and one at the occipital condyles. Model parameters were found by iterative adjustment of the parameters until the simulated results compared well with the experimental results of volunteer sled tests at a certain impact level. Simulating tests at other impact levels showed discrepancies between the numerical and experimental results. Results could be improved by relocation of the lower pivot posterior and inferior to T1,

resulting in a longer neck link with which a better decoupling of head angular and linear motion is obtained.

From the studies discussed above, it can be concluded that two-pivot models are indeed capable of simulating global head behaviour quite accurately. However, a two-pivot model suited for all impact directions and impact severities could not be obtained thus far.

4.2.2 Multi-Pivot Models

Recently, Daru [20] (as is mentioned in Ref. [119]) and Snijders *et al.* [106] developed three-dimensional quasi-static multi-pivot models of the cervical spine to study the effect of muscular action on head-neck kinematics. Rigid links, representing one or more vertebrae, are connected by pivot-joints to which torque-rotation characteristics are assigned, representing the behaviour of the vertebral segments. The cervical spine is not represented in detail, therefore these models are not discussed further.

4.3 Detailed Models

In *continuum mechanics models* the (cervical) spine is represented as a homogeneous bar or beam column). This beam may have varying dimensions but will have the same homogeneous material properties throughout. Thus, both geometry and mechanical behaviour are simplified. Analysis is performed using the principles of solid mechanics and (in the linear case) closed form solutions can be obtained. Continuum models are used, for example, to study the wave propagation phenomenon and are frequently encountered in studies on the pilot ejection problem. They permit a complete qualification of stresses and strains in the structure. However, discrete parameter models were developed to include the complex features of the spine.

In *discrete* or *lumped parameter models* the spine is considered as a structure formed by various anatomical elements, such as vertebrae and discs, to which different properties are assigned. In these models the spine is idealized as an assemblage of individual rigid vertebrae, connected by massless spring and damper elements representing the intervertebral disc and the surrounding soft tissue complex. Mass and inertial properties of the system are lumped into the rigid vertebrae. Muscles may be included too and, for example, represented by a passive spring, an idealized force generator or a dynamic muscle model.

Although, continuum models require a continuum representation of the characteristics of the spine, continuum mechanics and discrete parameter models may be equally viable. This is true if continuum models are refined to simulate more closely the anatomical function of the vertebral column. Yoganandan *et al.* [124] state that "if the discretization employed in the continuum models is similar to the discrete models, the governing continuum equations would be similar to that of the other type". However, King and Chou [61] already noted that "there is a tendency to seek analytical solutions of continuum formulations at the expense of a more realistic simulation".

Continuum mechanics and discrete parameter models cannot completely quantify the mechanics of the spine, because of the complex geometry, material inhomogeneities and the nonlinear mechanical response of the spine. To overcome (a part of) these limitations finite element models were advanced.

In *finite element models* the spine is also considered as a structure formed by various

anatomical components, but now each component is broken down into a large number of deformable elements which are in contact with each other. Each element has the continuum material properties of the anatomical component it belongs to. The mass is concentrated (lumped) at points in the corners or along the sides of the elements. In principle, the finite element method can accommodate any type of geometry, loading, material behaviour and boundary condition data. So, using the techniques of the finite element method enables it to give a detailed description of the mechanics of the structure.

The major drawback of finite element modelling is that it yields complex models (with many parameters) that are computationally inefficient compared with discrete parameter models. In fact, discrete models are a subset of finite element models: a subset of simplified models with fewer degrees of freedom and, therefore, fewer equations. This allows that nonlinearities are easier handled and that simpler solution methods may be employed.

Discrete parameter models are excellent, if the primary purpose is to represent the anatomic kinematics; however, these models cannot calculate stresses in the tissues or trace distribution of force in the various internal body structures [115]. To take advantage of both methods, 'hybrid' modelling can be used. In hybrid modelling, finite element representations are combined with discrete parameter models: "to reduce the complexity of the finite element representation, regions of the body that are remote from the area of interest are approximated with lumped parameter idealizations" [115]. Hybrid modelling has been used recently in crash-victim simulations [14,44].

4.3.1 Continuum Mechanics Models

No detailed continuum mechanics models of the (cervical) spine have been reported in the literature. Therefore, continuum mechanics models are not reviewed here, except for the recent, noteworthy models of Dai and Liu, and McElhaney and co-workers. For an elaborate description of continuum models, see Yoganandan *et al.* [124].

Liu and Dai [65] idealized the (straightened) cervical spine as a homogeneous isotropic beam column. Using the concept of stiffest and second stiffest axis, they determined analytically the loading conditions (magnitude and direction) for which compression failure or buckling failure of the cervical column occurs. They derived the safe load region for all possible directions of load applied to the column.

McElhaney and co-workers used the quasi-linear viscoelastic model originally proposed by Fung (e.g. [37]) to model the structural response of the cervical spine in compression [75], flexion [24] and torsion [82]. Model parameters were obtained from relaxation tests conducted on cervical spine specimen. Subsequently, the models have been validated for cyclic loading the cervical spine in compression, flexion or torsion. Myers *et al.* [82] used an improved method to obtain the model parameters, with which the responses of the cervical spine to cyclic torsional loading were successfully predicted. According to Myers *et al.*, the model performance for torsion is similar to the performance for flexion and compression, reported previously, suggesting that the model may be used as a model for the generalized gross-motion structural response of the cervical spine to loading.
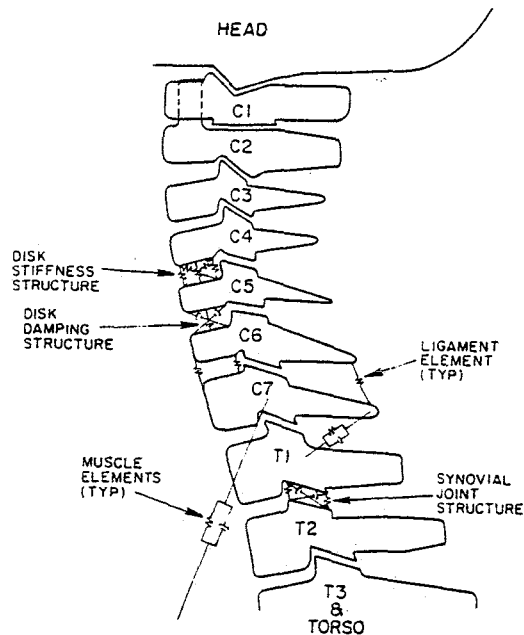
Figure 4.1: The model of Reber and Goldsmith.

4.3.2 Discrete Parameter Models

Two-Dimensional Models

Two-dimensional discrete parameter models are only briefly mentioned. A more extensive description of these models is provided by Yoganandan *et al.* [124].

Orne and Liu [87] were among the first to model the human spinal column as a discrete system consisting of individual vertebrae connected by intervertebral discs. In their two-dimensional discrete parameter model, vertebrae and head were modelled as rigid bodies of variable size and mass, and intervertebral discs as massless, deformable (visco)elastic elements. Their approach was used by McKenzie and Williams [77,117], who developed a detailed model of the head-neck system to study head-neck behaviour in frontal impacts. Melvin *et al.* [78] adapted the model of McKenzie and Williams to develop an improved dummy neck. They used the mathematical model to compare the responses of the dummy neck with those found in human volunteer tests. Prasad and King [97] advanced the Orne and Liu model by including facet elements as a secondary loading path along the spinal column. The facets are represented as springs connected to the vertebral bodies by a massless rigid rod. Tennyson and King [108] adapted the model of Prasad and King to simulate the effect of active spinal musculature on head, vertebrae and pelvis in vertical (ejection) and frontal impact accelerations. Pontius and Liu [96] used the modelling approach of Orne and Liu. They included (active) muscle-elements in a model of the cervical spine to study the effect of neuromusculature on head-neck motions during whiplash. In all models mentioned above, atlas and axis are modelled in the same manner as the other vertebrae, that is as rigid vertebrae connected by intervertebral discs.

Reber and Goldsmith [99] developed a detailed two-dimensional discrete parameter model to study the head-neck motion under whiplash conditions. The model included torso, vertebrae T2 through C1 and the head as rigid elements (Fig. 4.1). Intervertebral discs, ligaments and articular facets are represented in detail with complex spring-damper mod-

els. This model was extended into three dimensions by Merrill *et al.* (see section 4.3.2).

Recently, Li *et al.* [63] developed a two-dimensional quasi-static model of the cervical spine to determine the loading distribution imposed upon the cervical spine in extension-compression. It was developed to evaluate the hypothesis that extension is a risk reducing strategy in impact situations. Since the model is specific to this application (and two-dimensional) it will not be discussed further.

Three-Dimensional Quasistatic Models

Arvikar [4] developed a three-dimensional quasistatic model of the entire human musculoskeletal structure. The model is also described in Seireg and Arvikar [104]. The skeletal bones are modelled as rigid bodies connected by joints and muscles. Bones are kept in equilibrium by the tensile forces in the muscles. Here, only the modelling of the cervical spine will be discussed. All major muscles of the neck are modelled with multiple lines of action. The disc is simply modelled as a spring acting along the line connecting the middle of the endplates of two adjacent vertebrae. Other spinal elements are not included in the model.

Gracovetsky *et al.* [42] developed a three-dimensional model of the cervical spine and head. The model calculates the muscle forces needed to keep the head-neck system in its initial position when the system is subjected to forces resulting from sagittal plane loading conditions. The model includes skull and vertebrae C7-C1 as separate elements. Thoracic vertebrae T6-T1, sternum, ribs and shoulders are modelled as part of a fixed rigid base. The model includes a large number of muscle groups of which the points of insertion on the bony parts of the system are accurately determined. The model accounts for the forces due to the weight of head and neck and for the effect of external forces. These forces have to be balanced by forces generated by muscles, ligaments and intervertebral joints. It is a quasistatic model in that the initial configuration of head and neck is held the same when deceleration forces are applied. The model is not usable any more once the muscles have lost control of this fixed situation.

Three-Dimensional Dynamic Models

The first three-dimensional discrete parameter models of the spine were developed by Panjabi [90] and Belytschko *et al.* [9]. Both Panjabi and Belytschko *et al.* describe a general method for constructing three-dimensional discrete parameter models and the governing equations of motion, and both took the human spine to illustrate their methods. The elements used in the method are rigid bodies and deformable elements (springs and dampers). Of note, the method of Belytschko *et al.* was used by Schultz *et al.* [103] to develop a three-dimensional static model of the thoracolumbar spine.

Chen [18] developed a three-dimensional model of the human ligamentous spine suitable for use in both static and dynamic loading situations. Included are rigid vertebral bodies, deformable discs and posterior spinal elements (facet joints and ligaments), and the initial curvature of the spine. Chen reduced the model to two dimensions to analyze the pilot ejection problem.

Huston *et al.* [51,52] developed a head-neck model to predict head motion in impact situations. The model comprises nine rigid bodies, representing torso, cervical vertebrae and skull, connected by intervertebral discs, ligaments and muscles (Fig 4.2). Discs and muscles
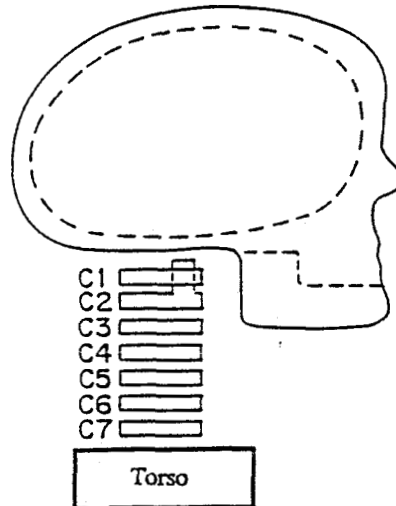
34

Figure 4.2: The model of Huston *et al.* and Tien and Huston.

are modelled as (visco)elastic solids, ligaments as nonlinear elastic bands. Both muscles and ligaments exert force only in tension. One-way dampers are used to model joint constraints which limit the relative motion of the bodies. Tien and Huston [109] simplified this model by taking an overall representation of the force-displacement behaviour of the soft tissue complex (disc, muscles and ligaments). They used empirical expressions for the forces and moments exerted by the soft tissues on the vertebrae. This yielded a computationally more efficient model with less parameters. Values for the (spring and damper) parameters were obtained from curve fitting of model prediction with experimental results of volunteer sled tests, and (hence ?) a good match between the numerical and experimental found head accelerations and velocities was obtained. The resulting model was further simplified by Tien and Huston [110], who fused the cervical vertebrae into a single rigid body, which resulted in the three-segment two-joint model mentioned in Section 4.2. To investigate the dynamic response of the jaw during whiplash (extension), Schneider *et al.* [102] added a moveable rigid jaw to the model of Tien and Huston [109]. The jaw-head joint allowed for both rotation and translation during jaw opening; jaw-motion was solely determined by the inertial characteristics of the jaw.

Suh [107] describes a method to construct a dynamic model of the cervical spine, which is based on the quasistatic model of Hong [45]. No quantitative data of the model are given, because not enough data on material properties and such were available at that time. Skull and vertebrae are modelled as rigid bodies. Ligaments and muscles in passive mode are modelled as nonlinear spring-dampers; and muscles in active mode as force generating elements. Facet joints are modelled with nonlinear spring-dampers that are compliant in tension and stiff in compression to allow easy movement of the desired type while resisting other types of movement. The modelling of discs is based on results of dynamic experiments on motion segments from which force-displacements characteristics are obtained for all independent loading directions. To simulate disc behaviour, it is assumed that the overall effect of a complex (combined) displacement is the sum of the independent displacements for which force-displacement characteristics were measured.

Merrill *et al.* [41,79] extended the Reber and Goldsmith model [99] into three dimensions. The resulting model was further improved by Deng and Goldsmith [22]. This lumped
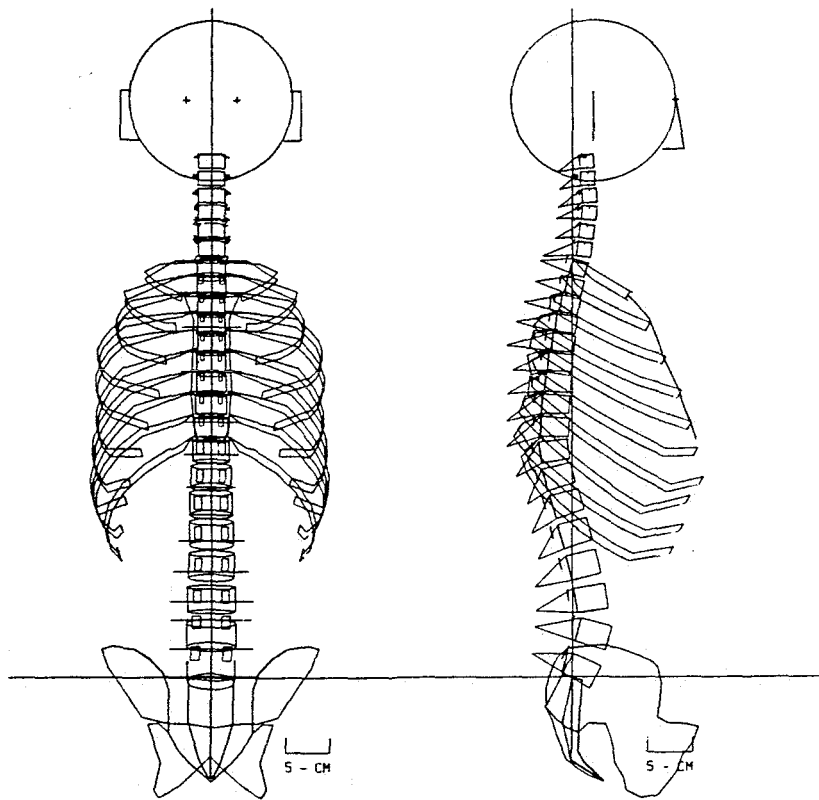
Figure 4.3: The model of Belytschko *et al.*

parameter model of head, neck and upper torso comprises ten rigid bodies representing torso with T2, the vertebrae T1 through C1, and the head. The overall mechanical response of intervertebral discs, ligaments and articular facets is lumped into a single force-deformation relation. The model also incorporates the major muscle groups of the neck, but only for the passive state. This model is discussed in detail in section 4.3.4. For the sake of completeness, the model of Luo and Goldsmith [69] is mentioned. They extended the model of Deng and Goldsmith to include the lower torso. The model comprises ten rigid bodies representing the head; the vertebral pairs C1-C2, C3-C4, C5-C6, C7-T1; the entire thorax; the lumbar vertebral combinations L1-L2, L3, L4-L5; and the pelvis.

4.3.3 Finite Element Models

Belytschko *et al.* [8, 7] developed a three-dimensional finite element model of the head-spine-torso structure to study the pilot ejection problem. The model includes the complete spine, pelvis and skull and may also include the rib cage and viscera (Fig. 4.3). Rigid vertebrae are connected by discs, ligaments and articular facets. These components are represented by several deformable elements which collectively provide resistance against axial, torsional, bending and shear loads. Although a more detailed representation of the neck is available within the model, only simulations with a simplified (beam element) representation of the cervical spine are reported. The model has been advanced recently by Privitzer and Kaleps [98] to study the effect of head-mounted systems on the dynamic response of the head and spine. Williams and Belytschko [118] used the approach of Belytschko *et al.* to develop a detailed head-neck model. The model comprises rigid vertebrae T1 through C1 and the skull connected by deformable elements representing
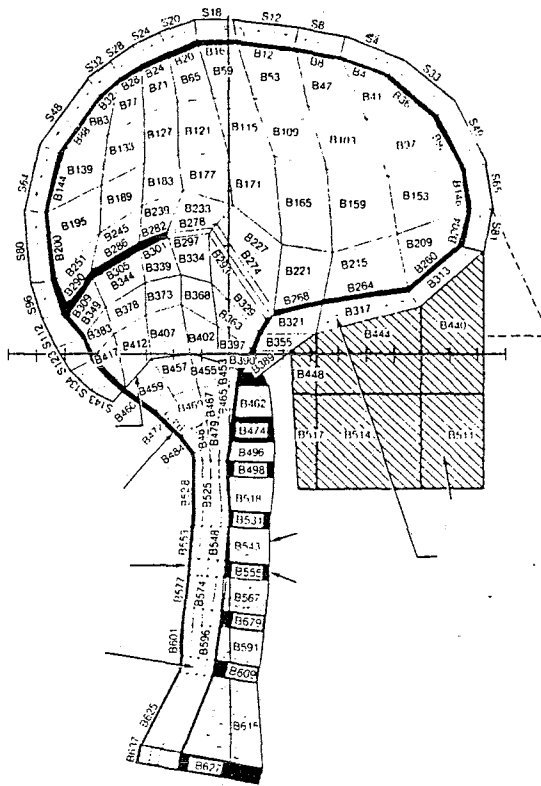
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Figure 4.4: The model of Hosey and Liu.

discs, facet joints, ligaments and muscles. This model is discussed in detail in section 4.3.5.

Hosey and Liu [46,47] developed a three-dimensional finite element model of the head and neck (Fig. 4.4). The model was developed primarily to study the mechanics of head (skull and brain) injury. It incorporates skull, dura, cerebrospinal fluid space, brain, jaw, cervical vertebrae and discs, and spinal cord. Each vertebra and each disc is modelled as a single element. Since the formulation is linear, the model is restricted to small displacements and rotations.

Dietrich et al. [23] described a three-dimensional finite element model of the human spinal system. The model includes the spine (vertebrae C3-L5), sacrum, pelvis and ribcage, modelled as rigid bodies. They omitted atlas and axis because of the different function and shape of these vertebrae. The soft tissue components are modelled with deformable finite elements. Both nucleus pulposus and anulus fibrosus of the intervertebral discs are modelled. Basic ligaments of the spine and important muscles that influence behaviour of the spinal system are included too. External forces (static load or inertial forces) can be applied to the model. The model allows for both static and dynamic analysis of forces occurring in the spinal system. An example of a static analysis is included in the paper.

The models of Hosey and Liu, Williams and Belytschko, and Belytschko et al. are discussed to a greater extent by Yoganandan et al. [124]. The lumped parameter model of Deng and Goldsmith and the finite element model of Williams and Belytschko seem to be the most advanced cervical spine models reported in the literature to date. Therefore, they will be discussed in detail in the following two sections.
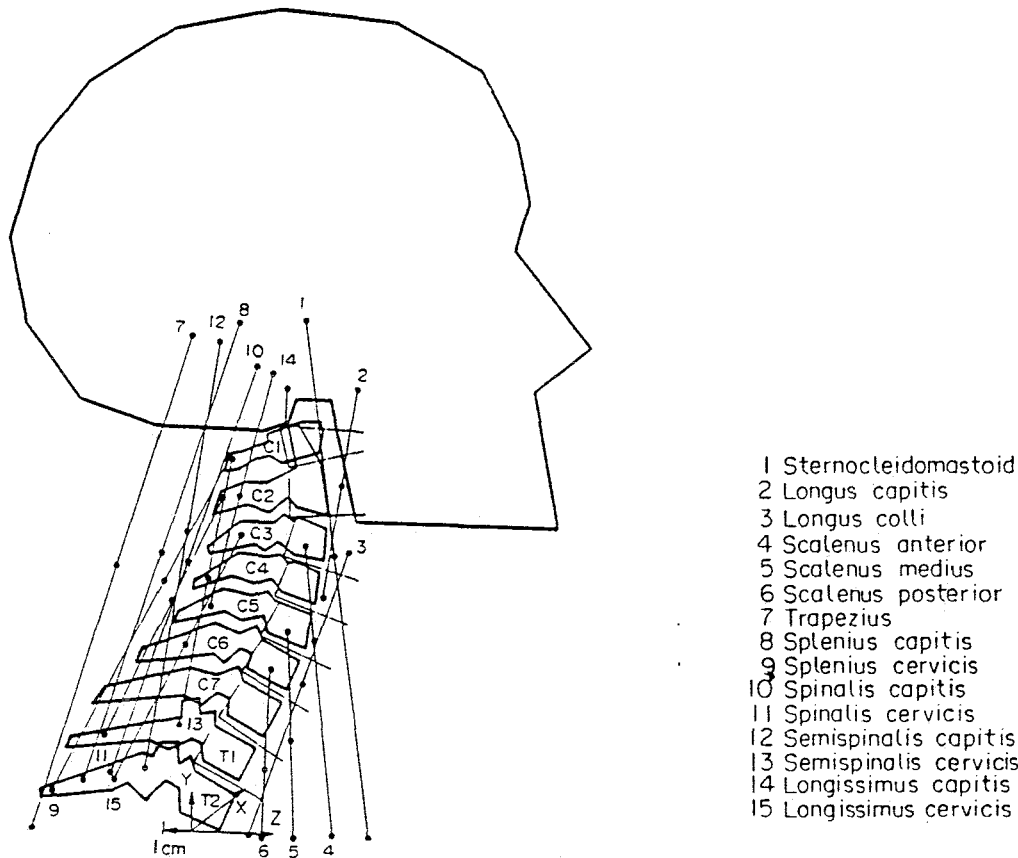
1	Sternocleidomastoid
2	Longus capitis
3	Longus colli
4	Scalenus anterior
5	Scalenus medius
6	Scalenus posterior
7	Trapezius
8	Splenius capitis
9	Splenius cervicis
10	Spinalis capitis
11	Spinalis cervicis
12	Semispinalis capitis
13	Semispinalis cervicis
14	Longissimus capitis
15	Longissimus cervicis

Figure 4.5: The model of Deng and Goldsmith. Configuration of the rigid bodies and muscle elements (with their names).

4.3.4 The Model of Deng and Goldsmith

Deng and Goldsmith (1987) [21,22] developed a three-dimensional *lumped parameter* model of the head-neck-upper torso to predict head-neck motion under impulsive loading and impact. Experimental results for frontal and lateral impact were used to validate the model. In this model, rigid bodies represent vertebra T2 with torso, vertebrae T1 through C1, and the skull (Fig. 4.5). The rigid bodies are connected by intervertebral joints, representing the mechanical behaviour of intervertebral discs, ligaments and facet joints. The model incorporates fifteen pairs of muscles of which only the passive state was modelled. This model is an improved version of the model developed by Merrill *et al.* [79,41].

The mechanical response of all intervertebral joints is lumped into a linear stiffness matrix, relating force and moment to translation and rotation. The off-diagonal elements of this matrix represent coupling of motion in one direction with load in another direction. Intervertebral damping is represented by a linear viscous damping term in the constitutive equation. Muscles are represented by three-point spring elements with nonlinear constitutive relationships. The use of midpoints allows the muscle elements to act along more realistic directions, than can be realized with the two-point elements used in the previous model [79]. Since only the passive state is modelled, active contraction of the muscles cannot be simulated.
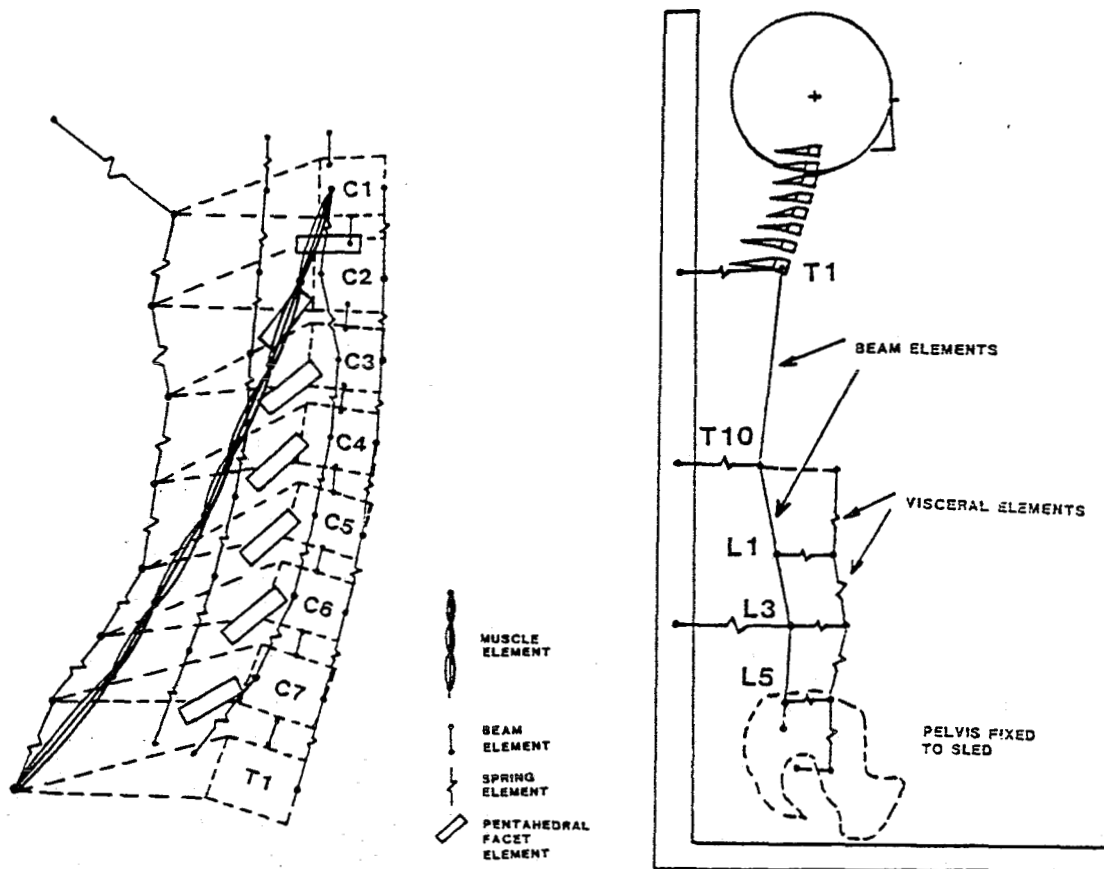
Figure 4.6: The model of Williams and Belytschko. (a) Rigid and deformable body representation of the cervical spine. (b) Cervical model connected to the simplified model of the lower spine and torso.

Validation To validate the model, numerical predicted head kinematics were compared to those from frontal and lateral volunteer sled acceleration tests of Ewing and Thomas [26] and Ewing *et al.* [31] respectively. Qualitatively, the response patterns were in reasonable agreement. Quantitatively however, correspondence is less good: especially the head accelerations remain well below those from the experiments during the initial impact phase. According to Deng and Goldsmith, discrepancies might be due to inappropriate values in the joint stiffness matrix, which were scaled from thoracolumbar spine data.

4.3.5 The Model of Williams and Belytschko

Williams and Belytschko (1983) [118] developed a three-dimensional *finite element* model of the human cervical spine for evaluating human response to impact situations. The model was validated for frontal and lateral impact accelerations. Simulations were performed to study the effects of stretch-reflex response of muscles on head-neck motion and on stress levels in the neck.

The head-neck model comprises nine rigid bodies, representing the vertebrae T1 through C1 and the head, which are connected by deformable elements, representing intervertebral discs, facet joints, ligaments and muscles (Fig. 4.6a). Twenty-two different neck muscle groups and a model for muscular contraction are included. They used the modelling approach of Belytschko *et al.* [8] which is based on the finite element method. The detailed

model is joined to a simplified model of the lower spine and torso described by Belytschko and Privitzer [10] (Fig. 4.6b).

Different deformable elements are used to model the intervertebral discs, facet joints, ligaments and muscles. Discs are represented by beam elements with linear torsional and bending stiffnesses and bilinear axial stiffness (that is, different stiffnesses in compression and in tension). Articular facets are represented by a special shaped continuum element, with axial and shear stiffnesses. Williams and Belytschko state that this facet element effectively maintains stability of the cervical spine in both lateral and frontal plane accelerations. Ligaments are represented by nonlinear spring elements, which have only axial stiffness. Muscles are represented by spring elements the axial force of which may be activated independently of the elongation to mimic muscle contraction. These elements include intermediate sliding nodes so that the muscles can curve around bones. Forces on the bones exerted by the curving muscles may be treated as well.

The intervertebral connections have the same structural arrangement for all levels from T1 to C2. Each vertebral pair is connected by one beam element and several spring elements. The arrangements between C2-C1 and C1-head are different to account for the unique properties of this region. The beam element between C1 and C2 allows for rotation of the atlas relative to the axis, but limits the amount of translation between C1 and C2. Two beam elements represent the synovial joints between the occipital condyles and C1.

Validation The model was validated by comparing the results of simulations to experimental data for frontal and lateral volunteer sled accelerations of Ewing and Thomas [26] and Ewing *et al.* [31] respectively. Simulations in which the muscles are passive throughout the simulation (passive muscle model) and in which the muscles start contracting after some time (stretch-reflex model) are compared to the experimental results. For frontal impacts, predicted head kinematics agree well the experimental results, whereby the model with muscle contraction gives slightly better results than the passive muscle model. For lateral impacts, correspondence is less good, showing substantial deviations between numerical and experimental (maximum) head acceleration and displacements.

4.4 Validation of Mathematical Models

In this section, experiments which can be used to validate a mathematical model of the human cervical spine are reviewed. A model is validated through comparison of numerical predicted results on head-neck responses to impacts with similar results obtained from experiments. A complete and thorough validation of a detailed model should include a comparison of results on both the global and the local dynamics and kinematics of the head-neck structure in various impact situations. Global refers to the forces on the head, neck and torso and to the motion of the head relative to the torso (T1). Local refers to the forces acting on each cervical component at each vertebral level and to the motion of each vertebra.

4.4.1 Global Validation

For global validation, the results of sled acceleration tests can be used. These tests have been performed with human volunteers and cadavers primarily to obtain the head-neck response to impact accelerations when direct head impact is not involved. Well known are the sled tests performed with volunteers at the Naval Biodynamics Laboratory (NBDL) in

New Orleans, Louisiana, and with cadavers at the University of Heidelberg in Germany. Results of these and other sled tests have been analyzed and are reported in the literature [11,26–30,32,55–57,71,120,121].

In a sled test, the subject (volunteer or cadaver) is placed in a seat. Motion of torso and upper and lower extremities is prevented by various straps and belts, but head and neck are not restrained. Accelerometers, mounted to the subject's head and to the torso at the level of T1, are used to obtain the overall dynamics (angular and linear accelerations) of the head and neck during impact. The sled is accelerated from zero velocity or decelerated from a certain velocity to cause inertial loading of the subject, resulting in motion of the head and neck. Some torso (T1) motion is seen too, since a complete restraint is not possible due to elasticity of the chest and straps. The overall kinematics of head and neck are obtained from high speed movies taken during the impact. In this way, sled experiments can give information about the global dynamics and kinematics of the head-neck system for various impact directions.

Volunteers have been subjected to moderate (non-injurious) impact levels, whereas cadavers have been subjected to moderate (for comparison with volunteer-tests) and severe (potentially injurious) impact levels. Accelerations may be applied in several directions: frontal, lateral, oblique, rear-end and vertical (pilot ejection). However, no results of experiments with rear-end impacts that give information about the global dynamics and kinematics have been reported in the literature. For volunteers this data are, indeed, difficult to obtain due to the vulnerability of the neck for rear-end impacts.

4.4.2 Local Validation

For local validation, a detailed knowledge of the dynamic and kinematic response of the human cervical spine is needed. This includes the kinematics of each bony segment and associated connective tissues along with the stress and strain throughout the soft tissue up to the level of anatomical disruption. Most of this knowledge cannot be experimentally determined from volunteers and is even hard to obtain from experiments with cadaveric material. As a consequence, only few experiments that may give such results have been reported in the literature.

Yoganandan, Pintar and co-workers [94,95,125,126] conducted experiments on head-neck specimen that give information about the kinematics of vertebrae when the cervical spine is subjected to axial compression. Both quasistatic and dynamic compressive loads were applied up to failure of the specimen, to study the injury biomechanics of the cervical spine. The cervical spine was aligned to remove the lordosis. The global dynamic response was obtained from load cells, placed on both ends of the specimen. The sagittal plane movements of vertebrae and base of the skull were obtained from film-recordings of retroreflective markers placed in the bony parts (C0-C7) of the specimen. Detailed results on the movements are given.

Alem et al. [1] conducted sub-injurious and injurious experiments with complete cadavers, in which the head of the cadavers were axially impacted. The sub-injurious impacts were aimed at generating kinematic and dynamic response to define the mechanical characteristics of the undamaged head-spine system in superior-inferior direction. Measurements included the acceleration of head and T1 as well as high speed movies of the total cadaver and high speed X-ray movies of the cervical spine. However, Alem et al. did not report any of the results of nor any comment on the X-ray movies or localized kinematics of the

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spine.

Studies that may give the localized movement of vertebrae in quasistatic, voluntary (muscle induced) motion of the head include the following. Moffat and Schultz [80] conducted an X-ray study on the motion of cervical vertebrae in voluntary flexion and extension movements. Van Mameren [113] conducted a similar and more thorough study to obtain motion patterns in the cervical spine in the sagittal plane. During voluntary flexion and extension, X-ray pictures were taken at a rate of 4 frames/second. Vertebral kinematics were analyzed to obtain characteristics for various motion parameters. Margulies *et al.* [72] used magnetic resonance imaging to measure the *in vivo* motion of the cervical spinal cord in human volunteers for stepwise flexion and extension of the neck. From these images vertebral movements can be obtained too.

4.5 Discussion

A number of sophisticated models describing the head-neck dynamics have been reported in the literature. These include the discrete parameter models of Tien and Huston [109] and Deng and Goldsmith [22] and the finite element model of Williams and Belytschko [118]. Although the model of Williams and Belytschko includes more details, the level of discretization is similar to the one employed in the Deng and Goldsmith model. For example, the disc modelling in both models is essentially the same. The main difference between those models is the facet joint element used by Williams and Belytschko, which cannot be replaced by discrete spring elements, as was pointed out by Williams and Belytschko.

The models mentioned above have been validated, but only for a small number of impact situations and only for global (head) kinematics. Model predictions have been compared to the results of volunteer sled tests. In general, the numerical and experimental response patterns showed good correspondence. Quantitatively, correspondence was less good and substantial deviations were reported. It should be noted that only a very limited amount of physical properties data on cervical (motion) segments was available at the time the models were developed. Properties were either tuned until the model showed reasonable behaviour [109,118] or estimated from data obtained from thoracolumbar components [22], in which case the model performance was less good. More accurate data may improve the model predictions.

When the performance of these models is compared to the performance of two-pivot models, it appears that the performances are about the same. Hence, detailed and pivot models seem to be equally capable of simulating the global behaviour of the head-neck system in various impact situations.

To date, more data have become available both on physical properties (see Section 2.3) and on experimental validation. For global validation, sufficient data are available on head-neck responses for various impact directions in which no head impact is involved, except for extension movements (rear-end impacts). Localized kinematics may be obtained from the results of axial impact tests, but these experiments cannot be compared to any of the sled tests. Within the sagittal plane, localized kinematics may be obtained from the quasistatic X-ray or MRI volunteer tests. Unfortunately, these tests are also not comparable to the sled tests since the latter are dynamic. Information on soft tissue forces (local dynamics) in both quasistatic and dynamic tests is not available at all.

Thus, detailed information is found only for the localized kinematics of the spine in axial compression (for quasistatic and dynamic loading) and in flexion/extension (for quasistatic loading). No results on localized dynamics have been reported to date. No detailed results of experiments comparable to the volunteer and cadaver sled tests have been reported either. Obviously, there is need for experimental results on the local kinematics and dynamics of the head-cervical spine structure in impact situations comparable to the sled tests performed with cadavers and volunteers. Results of these experiments together with those from sled tests may then serve as a database for validation of cervical spine models.

Another source for local validation might be the experiments conducted on cervical motion segments. A motion segment can be seen as a building block of the cervical spine. In quasistatic and dynamic experiments on motion segments, reported in the literature (see Section 2.3), segments of the upper and lower cervical spine have been subjected to well controlled loading and both loading and response have been carefully analyzed. The results of these experiments can be used to validate a mathematical model of a single motion segment of the lower cervical spine. The upper cervical spine can be treated similarly. Then, a model of the entire cervical spine can be built from these lower and upper cervical segment models. Finally, the performance of this model needs to be validated for the global dynamics and kinematics, to obtain a valid three-dimensional detailed dynamic model of the human cervical spine.

5 Conclusions

Goal of the project is the development of a mathematical model of the mechanical behaviour of the human cervical spine. The model must be able to describe the biomechanical response of the human head and neck to various impact situations (various directions and magnitudes of impact forces). Furthermore, the model must incorporate injury mechanisms: it must describe the local kinematics and dynamics of individual vertebrae and connecting soft tissues.

The functional anatomy and biomechanics of the cervical spine are described in Chapter 2. Qualitatively, the function of the biomechanical relevant components of the cervical spine is quite clear. Quantitatively, knowledge on the mechanical behaviour of cervical components and motion segments is still incomplete. The review on the physical properties data of the constituent components of the cervical spine, showed that for some of the components material characteristics are not available, while for other components data on the material characteristics are incomplete with respect to types (quasistatic, dynamic) and directions of loading.

Injury mechanisms and injuries of the cervical spine are discussed in Chapter 3. Injury mechanisms are described by the principal applied loading of motion segments of the cervical spine. Hence, injury mechanisms can be incorporated into a mathematical model if the model can describe the forces and deformations occurring at each vertebral level during the impact situation under study. Although in real-life accidents, the injury mechanisms are more complex due to unknown factors modifying the (principal) load applied to motion segments, the model can be used to simulate experiments to validate its injury predictability.

Mathematical models of the cervical spine and experimental data to validate these models, are reviewed in Chapter 4. In the literature, a number of detailed models as well as relatively simple two-pivot models describing the head-neck dynamics have been reported. The pivot and detailed models are equally capable of simulating the global head-neck motion in reasonable agreement with experimentally obtained results. Improvements can still be made here, however. Local vertebral movements can only be simulated by the detailed models, but these models have only been validated for global movements. For validation of the global behaviour of the model, sufficient data are available, except for rear end impacts. For validation of the local kinematic and dynamic behaviour, data are available for a few impact situations only.

In conclusion, two important problems in modelling the mechanical behaviour of the cervical spine are (1) the incompleteness of experimentally obtained data on physical properties, needed to develop a detailed model, and (2) the incompleteness of experimentally obtained data to validate a model.

Modelling Approach

In this project, the following approach to develop a detailed mathematical model of the mechanical behaviour of the human cervical spine is chosen.

First, a global discrete parameter model will be developed and validated, following Deng and Goldsmith. In this model, the behaviour of intervertebral soft tissue components is considered as a whole, that is, the components are not modelled separately but as one unit representing motion segment behaviour. Thus, only physical properties data describing motion segment behaviour is needed. This model will describe global head-neck dynamics and local vertebral kinematics.

Then, this model will be refined to obtain a model in which all relevant components are modelled separately. Models of these components have to be developed and parameters have to be estimated from experimental results, the availability of which is limited. The refined model will also describe the local dynamics of the components.

A refined model can be obtained three-fold: (1) refinement of the global discrete parameter model through replacing the intervertebral joint element by discrete (spring-damper like) elements representing individual components; (2) development of a complete finite element model; (3) hybrid modelling through replacing elements of the discrete parameter model that need to be modelled more precisely by finite element representations. The hybrid model combines the relative (numerical) simplicity of the discrete parameter modelling with the detailed modelling capability of the finite element method.

In another approach, the global discrete parameter model of the cervical spine will not be refined. Instead, a detailed (finite element) model of a cervical motion segment, which includes all relevant components, will be developed. The global model is used to obtain information on the global dynamics and local kinematics, which can be used in the motion segment model to study the mechanical behaviour of a cervical motion segment and its constituent parts in detail. Parameters of the detailed model can be obtained partially from the limited amount of physical properties data on components. The other parameters have to be tuned (estimated) from simulating experiments on motion segments. Estimation of the unknown parameters in a motion segment model is less difficult than in a cervical spine model, since a motion segment model has less parameters.

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