LOAD-SHARING IN THE HUMAN CERVICAL SPINE

by

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0-612-45264-6
To my two wonderful daughters

and their “booful mummy”

Thank you
Abstract

By virtue of the relative slenderness of the neck and the weight of the head, the cervical spine can be subjected to considerable loads. Knowledge about the basic biomechanics and load-sharing of the cervical spine may help to elucidate injury mechanisms or even suggest mechanisms leading to cervical joint degeneration. The aim of this study was to characterise the load transmission paths through cervical functional spinal units (FSUs). Compression and flexion, extension, lateral bending and torsional bending moments and anterior, posterior and lateral shear forces were applied. Under compression loading, the influence of superimposed flexion and extension postures was established. Under all other loading modes the effect of a constant axial preload was evaluated. The preload represented an active neck musculature and the weight of the head. An optoelectronic motion analysis system was used to measure specimen kinematics. Load-sharing mechanisms were studied using a miniature pressure sensor to measure intervertebral disc pressure and tri-axial strain gauges mounted beneath each of the facet joints and on the anterior surface of the vertebral body.

Our results suggest that flexion and compression-flexion loading modes result in compressive force in the anterior column and tensile (or small compressive) forces in the posterior column. Extension moments resulted in compressive force at the posterior column and tension in the anterior column. Extension-compression caused compressive forces in both the anterior and posterior columns. Lateral bending moments were associated with higher loads in the facet toward which the segment rotated. Torsion moments caused equal loading at each of the facets. Under anterior shear the vertebra pivoted about the compressed annulus anteriorly and ramped up the facet joints posteriorly resulting in a net distraction of the intervertebral disc and decrease in disc.
pressure. Posterior shear resulted in a pure posterior translation tension at the anterior annulus and distraction of the facet joints. Lateral shear caused an impingement of the facet joint on the underlying lamina. These results identify which specific anatomic structures may be highly loaded under each of the loads applied. Clinically, they can be used to identify parts of the anatomy that should be examined for injury.
Co-Authorship

This study was performed independently by the author, except that the experimental apparatuses were constructed by employees of the Müller Institute for biomechanics. The technique developed for animation of biomechanical tests (Appendix 1) relied in large part on pre-existing methods used by the Computer Aided Surgery research group at the Müller Institute. In particular, the work relied on software developed by two co-authors on that appendix, Marwan Sati and Yvan Bourquin.

The comparison of experimental application of preload (Appendix 5) was conceived and designed by the author. In addition the author had previously developed one of the preload application jigs. For the work presented in Appendix 5, Sabina Bruehlmann, a research engineer at the Müller Institute, performed the day-to-day experimental work, much of the analysis, and some of the manuscript preparation under the author's supervision. Also for Appendix 5, Drs. Orr, Oxland and Nolte provided feedback.
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CHAPTER 1. GENERAL INTRODUCTION

1.1. Overview of relevant cervical spine anatomy and physiology

The cervical spine is the most flexible region of the human spine. In addition to allowing this flexibility, it must support the head weight, the considerable forces that can develop, because of head or body acceleration and the traumatic forces inherent to head or neck impact. These functions must be accomplished while protecting neural and vascular tissues going to and returning from the skull which are vital for survival.

Although neck injuries account for a relatively small percentage of all serious trauma (2-4%),\(^1\) they are frequently associated with death or paralysis. Interest in neck biomechanics has therefore traditionally focused on mechanisms of traumatic injury, devices to prevent injury, and the efficacy of clinical – especially surgical – procedures. Over the last decade, the incidence of “whiplash”-associated injuries and symptoms – and the associated cost to society - has risen.\(^2\) As a result, interest in the biomechanics of whiplash injury has increased. Degeneration in the cervical spine is also common.\(^3\) In extreme cases it can lead to compression of spinal nerves or the spinal cord; each of which is manifested clinically as pain and other debilitating symptoms.

Biomechanical engineers apply the techniques of mechanical engineering to the study of human musculoskeletal and other biological systems. These studies are often conducted in conjunction with other physical and medical scientists and clinicians. To aid in communication, the standardised terminology to describe transectional planes and relative directions with respect to the human body will be used in this thesis (Figure 1-1).
The cervical spine consists of seven vertebrae (C1-C7, Figure 1-2). It articulates caudally with the first thoracic vertebra and supports the head cranially. The uppermost two vertebrae are each somewhat atypical compared to the lower five. The uppermost vertebra, C1, supports the skull atop a pair of facet joints, which articulate with occipital
condyles at the skull’s base. The C1 vertebra is also called the atlas; it is ring-like with two relatively large lateral masses that transmit load from the skull to lower vertebrae.

![Figure 1-2: Cervical spine anatomy (after Netter)](image)

The second vertebra, C2, has a central vertebral body with a protruding bony column: the dens. This vertebra, also referred to as the axis, articulates with the atlas superiorly via two large articular facets and there is an articulation between the dens and anterior ring of the atlas. The axis articulates inferiorly through two smaller and more laterally located facet joints and the two vertebral bodies articulate centrally through an intervertebral disc (IVD).

The rest of the cervical vertebrae (C3-C7) are relatively similar to each other. They have central vertebral bodies connected to each other by IVDs and articulating at bilaterally and posteriorly located facet joints. Pairs of cranial and caudal facet joints are connected by bony structures called lateral masses. Each vertebra has transverse processes to which muscles attach and which provide protection for blood vessels running through them. Of these, the vertebral artery is especially important because it
supplies blood to the brain. The so-called posterior arch consists of the spinous processes and other bony elements (laminae and pedicles) which connect these to the vertebral body.

The vertebral body and posterior arch together form the vertebral foramen through which the spinal cord passes. Adjacent vertebrae then form a protected bony vertebral canal. Spinal nerves run laterally from the spinal canal through the intervertebral foramen. Each of the vertebrae C3 to C7 have small bony protuberances (uncinate processes) projecting superiorly from the lateral margins of their cranial endplates. The cervical spine is concave posteriorly in its neutral position. This concavity is referred to as cervical lordosis.

The basic motion unit of the spine is a so-called functional spinal unit (FSU) consisting of two adjacent vertebrae and the IVD and other intervertebral tissues which connect them. The major intervertebral structures present in the cervical spine are depicted in Figure 1-3. The intervertebral disc and vertebral bodies are often referred to as the anterior column of the spine and the lateral masses and posterior arch are referred to as the posterior column.
The cervical IVD consists of outer layers of fibrous tissue, together called the annulus fibrosus (Figure 1-4). At the centre of the disc, the annulus gradually gives way to a gelatinous structure, the nucleus pulposus, which has a high water content.
Figure 1-4: Structure of the cervical IVD (after Mercer and Bogduk\textsuperscript{15}). The nucleus (np) is contained by the layers of the annulus (a). The annulus is thickest anteriorly.

1.2. \textit{In vitro spine testing}

One challenge of spinal biomechanics is that the \textit{in vivo} loading situation cannot usually be directly measured which means that our understanding of the loading situation depends on mathematical models which are also subject to limitations.\textsuperscript{13} Many investigators conform to the paradigm of applying isolated forces or moments \textit{in vitro} first advocated by Panjabi.\textsuperscript{22} Under this paradigm isolated bending moments or forces are applied to one vertebra of an FSU while the other vertebra is fixed. The free vertebra is free to move in response to the applied load. The motion that results from the applied load is measured and typically used to calculate specimen stiffnesses (load/displacement) or “flexibilities” (displacement/load). This is often called flexibility testing and the dependant variable is usually the applied load.

It is well known that the \textit{in vivo} situation is more complicated than the isolated loads applied \textit{in vitro}.\textsuperscript{29} However, the flexibility testing approach has the advantage that it provides repeatable data which can be reproduced at different laboratories. It also minimises the effect that the inter- and intra- specimen variability inherent to biological specimens has on the measured variables.
The fidelity between in vitro tests and the in vivo situation can be improved by applying a so-called compressive preload. This simulates muscle activity and the weight of the body cranial to the spine segment. The preload compresses the joint and remains constant throughout the test. Compressive force has been identified as the most important load component differentiating the in vitro and in vivo situations.29

1.3. Load-sharing in the human cervical spine

The investigation of cervical spine load-sharing mechanisms involves analysing the way that applied loads are distributed between relevant anatomical components of the spine. Applied loads include compressive forces, shear forces and bending moments. The anatomical structures that support applied loads include the vertebrae, spinal ligaments, joint capsules, facet joints, IVD and the neck musculature. Load-sharing analyses attempt to determine how the applied loads are distributed among the anatomical structures. Load-sharing information has both clinical and basic applications. Clinically, the tissues known to be the most highly loaded under a particular traumatic load can be specifically examined for injury. Likewise, load-sharing information may suggest previously unsuspected injury mechanisms for specific clinical symptoms. This is especially relevant for injuries for which organic lesions are suspected but are too small to be detected using contemporary diagnostic techniques. The aetiology of segment degeneration and instability may also depend on mechanical factors such as load magnitudes and load-sharing mechanisms.10

Mathematical models allow the investigation of loading scenarios or of physical variables that are impractical or impossible to evaluate otherwise.23 Several models of the cervical spine have been developed and they have been used to focus on many different questions.2,5,7,11,16,36,39 Model complexity increases with realism and often depends to a large extent on the available computing power. Accordingly, contemporary models have become very complex and potentially very powerful. Models must be
validated against experimental data before the model results can be considered valid.\textsuperscript{11}

The validity of a model increases with the number of different loading scenarios and measured parameters that are available for validation. Unfortunately many experimental cervical spine studies have focused exclusively on specimen kinematics and other kinds of data (i.e. disc pressure or load-sharing data) is scarce. This limits the validity of contemporary cervical spine models. Maurel \textit{et al.}\textsuperscript{11} have referred to this:

"Because of the shortage of published data, further experimental studies are required to be able to introduce more precise mechanical characteristics for the ligaments and the intervertebral disc in order to be able to analyse more precisely local phenomena, like stresses in the disc. Indeed, some choices of modelling were made in this model which gave a good global biomechanical behaviour but other kinds of modelling could perhaps have done similar results..."

Experimental studies, which focus on load-sharing and measure force or strain parameters in addition to kinematics, will provide a needed resource for mathematical validation.

Many groups have investigated load-sharing or isolated component loads in the lumbar spine. In this region, pressure sensors have been used to analyse intervertebral disc biomechanics,\textsuperscript{14,18,31,33,35} strain gauges have been used to measure loads at the facet joints,\textsuperscript{30} vertebral body\textsuperscript{9,38} and endplate.\textsuperscript{3} Mathematical models have also been used to focus on load-sharing in this region.\textsuperscript{6,38} In contrast, there is almost no basic experimental data on cervical spine disc pressure behaviour.\textsuperscript{6,27} There are also very few studies, which have focused on load-sharing.\textsuperscript{5,21,25}

Disc pressure is normally measured by placing a small pressure sensor in the nucleus pulposus, which is the most liquid-like substance in the disc. The pressure sensor is typically incorporated in a needle that is used to penetrate the annulus and place the sensor in the nucleus. Hattori \textit{et al.}\textsuperscript{6} have measured the \textit{in vivo} disc pressure using a cohort of patients diagnosed with disorders of the cervical spine. Their study focused on the effects of posture and degeneration on disc pressure. Pospiech \textit{et al.}\textsuperscript{27}
investigated the effect of muscle action and ventral fusion on cervical disc pressure using whole cervical spines undergoing small moment application *in vitro*.

The reason for the scarcity of disc pressure for the cervical spine is likely associated with the size of cervical discs. They have much smaller dimensions than their counterparts in the thoracic and lumbar spine\(^4,26\) making placement of the pressure sensor in the nucleus difficult. Additionally, even very small diameter needle-mounted pressure sensors are large compared to the disc height.\(^4\) Both of the above mentioned studies used needle-mounted sensors. It is reasonable to suspect that the specimen biomechanics could be altered by the presence of a needle (even a very small one – Pospiech *et al.*\(^27\) used a 1.3 mm diameter needle) in the cervical region.

Pal and Sherk\(^21\) used load cells mounted separately under the anterior and posterior columns of the spine to estimate the distribution of force under compressive load. However, they applied very small loads (approximately 26 N) compared to that believed to be present *in vivo* (100-1000 N).\(^15\)

Pintar *et al.*\(^25\) estimated load-sharing in the cervical spine using uniaxial strain gauges mounted on the vertebral body and lateral masses. They applied pure and eccentric (producing flexion and extension moments) compressive loads to cervical spine segments. The use of uniaxial strain gauges by these authors implies that strains not aligned with the strain gauges (i.e. due to shear force application or tension in obliquely oriented ligaments or annulus fibres) would not be detected.

Goel and Clausen\(^5\) used a finite element model to analyse load-sharing in a C5-6 motion segment. The model predicted many important parameters such as disc pressure, facet joint force and ligament deformation. Because of the lack of experimental data of this type, their model could only be validated against the available kinematic data.
Some studies have focused on the effect of transection of ligaments or other structures. This approach certainly provides information about the contribution each component makes to segment stability. As such, indirect information can be gleaned about the probable role of the relevant structures for load-sharing. The approach is somewhat limited by the fact that it fundamentally depends on information from a pathologically altered or injured segment but seeks to make conclusions about healthy load-sharing.

1.4. Motivation

The main goal of this work was to analyse the load-sharing mechanisms in the human cervical spine under a clinically-relevant loading regime. The following limitations in the current state of knowledge about cervical spine biomechanics will be addressed:

i. Compared to the situation for the lumbar spine, there appears to be a paucity of data detailing the load-sharing behaviour in the cervical spine under moment, shear and compressive load application.

ii. Almost no basic information is available about the basic biomechanics of the cervical IVD. Contemporary techniques to measure intradiscal pressure may be inappropriate for use in the cervical spine.

iii. Tri-axial strain gauge rosettes have not previously been applied to cervical vertebrae to measure bone strains. These sensors have the advantage that the maximum bone strain present can be identified regardless of its orientation.

iv. The influence of compressive preload, representative of neck muscle activation, on cervical spine biomechanics has not been systematically investigated.

v. Kinematic data from biomechanical tests is usually presented in terms of ordered angles (i.e. Euler angles). Especially for spine kinematics, which can exhibit significant coupled motions, this format of data presentation is at best "non-
intuitive”. It presents unnecessary barriers to communication within the interdisciplinary team normally undertaking orthopaedic biomechanical research.

1.5. Objectives

The main objective of this study was to identify load-sharing mechanisms of the cervical spine under clinically-relevant isolated loading modes. In support of this objective, techniques to measure intervertebral disc pressure, bone strains and to interpret specimen kinematics were developed. A priority was to use loading apparatuses that applied isolated loads. Extraneous apparatus-related loads were minimised.

A quantitative analysis of the load shared by each anatomical element (i.e. individual ligament forces or pressure distributions across the IVD or facet joints) was beyond the capabilities of the methods applied in this study. The aim was to allow a qualitative analysis of the loads transmitted through the anterior and posterior columns of the spine.

1.6. Research outline

This thesis is organised using the manuscript format. Chapters 2, 3, 4 and 5 and Appendix 5 are articles. Chapter 2 and Appendix 5 have been submitted to the Journal of Biomechanics.

The study is summarised below. It consisted of three in vitro studies (Chapter 3-5) using a common set of cervical spine FSUs instrumented with load-sharing sensors. The three studies are differentiated by virtue of the loads applied. Compression with superimposed flexion and extension postures (Chapter 3), bending moments (chapter 4) and shear forces (Chapter 5) were applied.

Three techniques were developed in support of the load-sharing study. A novel technique was developed to measure intradiscal pressure in the cervical spine (Chapter
2). In two other parts of the study techniques to apply compressive preload (Appendix 5), and to animate kinematic results (Appendix 1) were developed.

1.6.1. A MINIMALLY DISRUPTIVE TECHNIQUE FOR MEASURING INTERVERTEBRAL DISC PRESSURE IN VITRO (CHAPTER 2, TECHNICAL NOTE)

A novel technique to measure intradiscal pressure is presented in Chapter 2. In order to minimise disruption to the specimen biomechanics, a technique was developed to insert a miniature pressure transducer into the nucleus pulposus through a tube which was subsequently removed. In contrast to needle-mounted sensors, only the miniature sensor (in the nucleus) and the three fine lead wires (passing through the annulus) were left in place during testing using this technique.

1.6.2. COMPRESSION FORCE TRANSMISSION IN THE HUMAN CERVICAL SPINE: EFFECT OF FLEXION AND EXTENSION POSTURES (CHAPTER 3)

A study examining the load-sharing mechanisms under non-destructive compression loading is presented in Chapter 3. Each FSU was instrumented to identify load transmission paths with an IVD pressure sensor and with tri-axial strain rosettes. The rosettes were used to measure bone strains and strain orientations at the vertebral body anteriorly and at the lateral masses posteriorly. The variation in load-sharing mechanisms as a function of physiologic flexion and extension postures was also investigated.

1.6.3. LOAD-SHARING AND HELICAL AXES OF MOTION IN THE CERVICAL SPINE (CHAPTER 4)

An investigation of load-sharing under flexion, extension, lateral bending and axial torsion bending moments is presented in Chapter 4. The effect of superimposed compressive preload on the specimen biomechanics was also investigated. The load-sharing sensors described above were used. The helical axis of motion (HAM) was calculated and used to aid in the interpretation of load-sharing mechanisms. The specimen kinematics were also analysed using the animation technique (Appendix 1).
1.6.4. Biomechanics of the human cervical spine under shear loading: influence of superimposed axial compression (Chapter 5)

Load-sharing was investigated under anterior, posterior and lateral shear (Chapter 5). The effect of superimposed compressive preload on the specimen biomechanics was evaluated. The same load-sharing sensors described above were used. Specimen kinematics were analysed using the animation technique (Appendix 1).

1.6.5. Animation of in vitro biomechanical tests (Appendix 1)

A novel technique to animate the kinematic results from biomechanical tests is presented in Appendix 1. This technique uses reconstructed CT models of the actual specimen tested to graphically replay three-dimensional kinematics. Behaviour of the HAM was also animated. Kinematic results could be presented in a much more intuitive manner than the usual graphs of ordered orientation angles.

1.6.6. In vitro axial preload application during spine flexibility testing: towards reduced apparatus-related artefacts (Appendix 5)

A quantitative comparison of four contemporary methods of compressive preload application is presented in Appendix 5. Some methods of preload application produced large extraneous loads. These loads were artefacts of the axial preload and the associated preload application technique. Recommendations to prevent the creation of preload-associated artefact loads were made.

1.7. References


CHAPTER 2. A MINIMALLY DISRUPTIVE TECHNIQUE FOR MEASURING INTERVERTEBRAL DISC PRESSURE IN VITRO: APPLICATION TO THE CERVICAL SPINE

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2.1. Abstract

A novel technique to measure in vitro disc pressures in human cervical spine specimens was developed. A miniature pressure transducer was used and an insertion technique was designed to minimise artefacts due to insertion. The technique was used to measure the intradiscal pressure in cervical spines loaded in pure axial compression. The resulting pressure varied linearly with the applied compressive force with coefficients of determination (r\textsuperscript{2}) greater than 0.99 for each of the four specimens. Peak pressures of between 2.4 and 3.5 MPa were recorded under 800 N of compression.

2.2. Introduction

The pressure in the nucleus pulposus of lumbar intervertebral discs has been extensively studied in both \textit{in vivo}\textsuperscript{10,11} and \textit{in vitro}\textsuperscript{1,2,7,9,15} settings. In this region of the spine, disc pressures have been shown to increase linearly with applied compressive load and to be hydrostatic in healthy discs.\textsuperscript{9} The effects of various physiologic and injurious loads and disc degeneration have also been extensively investigated.\textsuperscript{1,7,9,14,15} Very few studies have been undertaken in the cervical spine, despite the over three
decades of experience in the lumbar spine and the possible role of disc pressure in the mechanics of cervical disc disease and herniation leading to radiculopathy of the arm.

Hattori et al.⁴ and Pospiech et al.¹³ have measured in vivo and in vitro pressures respectively in cervical spines. The study by Hattori et al.⁴ focused on the effects of posture and degeneration on disc pressure. Pospiech et al.¹³ examined pressure in whole cervical spines undergoing small applied moments and determined the effects of simulated muscle action and ventral fusion. The variation in pressure with basic parameters such as applied axial compression has not yet been examined.

One reason for the paucity of cervical disc pressure data is the technical challenge associated with placing a pressure sensor in the nucleus pulposus. The uncinate processes prevent needle-mounted sensor insertion from lateral, and the inferior protuberance of the antero-inferior rim of the superior vertebral body often prevents or complicates insertion from anterior. The discs in this region have smaller cross-sections, both overall and of the nucleus, than those from other regions of the spine¹² making placement in the nucleus difficult to accomplish. Investigators to date have used needle mounted pressure sensors. Pospiech et al.¹³ used a 1.3 mm diameter needle and Hattori et al.⁴ did not document details with respect to needle diameter or insertion orientation. It is possible that needle mounted sensors alter the natural biomechanics of the segments in which they are implanted. The needle may restrict movement of the segment as a result of contact with the aforementioned bony structures. This in turn could cause needle deformation or sensor contact with the endplate either of which could induce an artefact in the pressure signal. The direct displacement of portions of the annulus and nucleus associated with needle placement could also alter the segment biomechanics.

The objectives of this study were to develop a technique for measuring cervical disc pressure in vitro which altered the natural cervical disc and bony anatomy as little
as possible in order to minimise distortion of the specimen's natural biomechanics. A related objective was to test the hypothesis that cervical disc pressure varies linearly with applied compression.

2.3. Materials and Methods

A guide tube was constructed from a 2.2 mm outer diameter, 1.4 mm inner diameter stainless steel needle with a sharpened tip. The tube was modified by first drilling out the inner diameter to 1.8 mm and then deforming it in a mechanic's vice such that an elliptical cross section resulted. Outer major and minor diameters of 2.45 mm and 1.70 mm respectively, and inner major and minor diameters of 1.75 mm and 1.10 mm respectively resulted (Figure 2-1 A). The guide tube was inserted into the disc from an antero-lateral direction, passing between the uncinate process and inferior protuberance (Figure 2-2 A and B). An adjacent endplate was used as a template from which the appropriate orientation and depth of insertion could be determined such that the sensor was placed in the nucleus at the centre of the disc (Figure 2-1 A).

The miniature pressure sensor (model 060 S, pressure range=0-3.5 MPa, Precision Measurement Company, Ann Arbor USA) was disc shaped and had one active measuring face (Figure 2-1 B). The deformation of a 350 Ohm strain gauge mounted inside the sensor was linearly proportional to the pressure acting on the measuring face. The strain gauge was connected as the active arm in a Wheatstone bridge circuit (a so-called \( \frac{1}{4} \) bridge). Temperature compensation was accomplished by connecting the sensor using a so-called "three-wire" circuit.\(^5\) The system was calibrated to 1.6 MPa in a pressure chamber after which output voltage from the amplified Wheatstone bridge was a known linear function of the applied pressure.

The sensor was inserted into the disc through the guide tube. An insertion wire (diameter=0.8 mm, Figure 2-1 A) was used to push the sensor into the disc nucleus (Figure 2-2 C and D). The insertion wire had a groove machined in one end, which fit
onto the pressure sensor disc between the two wire leads. Finally, the guide tube was withdrawn over the insertion wire and sensor cables (Figure 2-2 E), and then the insertion wire was withdrawn, leaving the miniature sensor embedded in the nucleus with only three 0.26 mm diameter flexible electrical cables passing through the annulus (Figure 2-2 F). Sensor position was validated using radiographs taken from anterior, lateral and superior directions. The opening made in the annulus by the guide tube was glued closed using a drop of “tissue glue” commonly used to close wounds (Histoacryl, Braun, Melsungen, Germany).

The technique was used to measure the pressure in four cervical discs from human functional spinal unit (FSU) preparations (levels: 1xC3-4, 1xC4-5 and 2xC2-3). The specimens were intact including intervertebral ligaments and posterior elements. Disc pressures were recorded as the specimens were subjected to compression in a materials testing machine. The specimens were compressed at a rate of 10 N/s under force control, to 800 N. Two pre-conditioning loading cycles were applied and data was collected from a third cycle. A linear regression was performed on the resulting data to test the hypothesis that cervical disc pressure varies linearly with applied compression.

2.4. Results

In all cases the post-insertion radiographs confirmed sensor placement within 2.0 mm of the geometric centre of the disc. The resulting pressure profiles were highly linear with coefficients of determination ($r^2$) greater than 0.99 (Figure 2-3). Peak pressures at 800 N ranged from 2.4 MPa to 3.5 MPa. In no case was nucleus or other disc material extruded through the hole made during sensor insertion.

2.5. Discussion

The technique presented represents an approach to measure disc pressure, which requires a minimal intrusion to the disc nucleus and annulus volumes. We believe
distortion of the pressure signals or specimen kinematic behaviour due to sensor insertion to be minimised using this approach. Three fine, flexible cables pass through the antero-lateral annulus compared to stiff circular needles, commonly greater than 1.0 mm in diameter, for needle mounted sensors used in lumbar disc pressure measurement. The effect of a rigid needle may well be magnified in the cervical spine where disc heights as small as 3.8 mm have been measured in a cohort of young men at the anterior margins of the disc. A 1.0 mm diameter pressure needle would represent 26% of this cervical disc height. Placement of the sensor using an adjacent endplate of the FSU for depth and orientation guidance was accurate in the four cases reported here.

Using this technique, it was not possible to control the orientation of the sensor’s measuring face. We assumed that the pressure in the nucleus was hydrostatic. Nucleus pressure is known to be hydrostatic in the morphologically similar lumbar IVD nucleus. In this case, the disc pressure is not dependent on the sensor orientation. Further work is necessary to confirm that disc pressure is hydrostatic in the cervical spine.

The disc pressure varied linearly with applied axial compression. This behaviour is similar to that observed for lumbar discs. The pressures measured are however, considerably higher than those reported by investigators who have tested lumbar spine specimens at comparable load levels. For example, Nachemson reported a pressure of 0.67 and 0.82 MPa using L3-4 and L2-3 discs respectively under 800 N of compression. McNally and Adams reported a pressure of 0.5 MPa for an L2-3 disc under 500 N of compression. This phenomena is not surprising since disc pressure for a given force is known to be inversely proportional to disc cross-sectional area. In vivo compression forces are also considerably smaller in the cervical spine as opposed to the lumbar spine. Gross “rules of thumb” for lumbar and cervical disc pressure can be estimated from this data. In lumbar discs approximately 1.0 MPa is induced for every 1000 N of
applied compression and in cervical discs approximately 3.75 MPa is induced per 1000 N of compression (based on an extrapolation of our data). More specimens must be measured using this technique in order to provide a “baseline” of data about cervical disc pressure.

2.6. References


2.7. Figures

Figure 2-1: A - Insertion apparatus. 1-specimen (level C4-S, viewed from antero-lateral and slightly superior), 2-insertion wire, 3-guide tube, 4-sensor. The cranial endplate of C4 was used as a “template” as shown (dotted line) from which the guide tube orientation and depth could be determined so that the sensor implanted at the centre of the disc (point 5). B - Close up of the pressure sensor with dimensions.
Figure 2-2: Insertion procedure using a C2-3 specimen with the dens removed. All views are from anterior or antero-lateral and slightly superior. A, B - The guide tube was inserted from antero-lateral. C - The sensor was mounted to the insertion wire. The machined notch fit onto the pressure sensor disc between the two lead wires. D - The sensor was pushed through the guide tube, to the centre of the disc, with the insertion wire. E - The sensor was held at the centre of the disc with the insertion wire while the guide tube was withdrawn. F - Finally, the insertion wire was withdrawn leaving only the lead wires passing through the annulus. The self-tapping screws visible in the figure were used to reinforce the connection between the specimen and the moulding material.
Figure 2-3: Disc pressure and associated linear regression fits, plotted as a function of applied compression.
CHAPTER 3. COMPRESSIVE FORCE TRANSMISSION IN THE HUMAN CERVICAL SPINE: EFFECT OF FLEXION AND EXTENSION POSTURES

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3.1. Abstract

Axial compression is a common injury mechanism in many traumatic injuries to the cervical spine and may play a role in whiplash injury. There is a paucity of basic data detailing how compressive and other loads are shared among the anatomic structures of the cervical spine. This information would allow confirmation of cervical spine injury models and may shed light on mechanical factors that are related to joint disease development. The objective of this study was to characterise the load transmission paths through cervical functional spinal units (FSUs) as a function of flexion and extension postures.

Six cervical FSUs were subjected to non-destructive compressive forces of 800 N at postures varying from full extension to full flexion at 2 degree intervals. Anterior-posterior translations were allowed during the compression. Specimen kinematics and inferior reaction loads were measured using an optoelectronic motion analysis system and a six-axis load cell respectively.
To identify load transmission paths, triaxial strain gauge rosettes were glued to the anterior surface of the caudal vertebral body and beneath the left and right facet joints on the lateral masses. A miniature pressure transducer was inserted into the centre of the intervertebral disc.

The anterior column transmitted compressive force even at high extension postures and the compressive force increased with flexion. The lateral masses were often unloaded or loaded in tension as flexion increased and supported increasing magnitudes of compressive force as extension increased. Disc pressure appeared to be invariant with changes in posture for these postures. Average disc pressures at 800 N of applied load were between approximately 2.4 and 2.7 MPa. The average disc compression at 800 N of applied load in the neutral posture was 0.83 mm. The maximum reaction load was a 10 Nm extension moment, which occurred at the maximum extension posture.

Our results suggest that the anterior column of the cervical spine transmits the bulk of the applied compressive force at all flexion and extension postures tested. At large extension angles, the posterior column also transmits compressive force.

3.2. Introduction

Axial compression is a common mechanism in traumatic injuries to the cervical spine.\textsuperscript{13} It has also been suggested that axial compression secondary to muscle activation could play a role in whiplash injury.\textsuperscript{12} Basic data detailing the manner in which axial compression is shared between anatomic elements of the cervical spine may help elucidate mechanisms of cervical spine injury. Previously unsuspected mechanical factors that may lead to joint degeneration or other disease may also be identified from this data. This information has applications clinically since it identifies anatomic elements that are highly loaded under a particular loading mode. These structures can be preferentially examined in patients known to have experienced traumatic instances of
the corresponding loading mode. Load-sharing results for the cervical spine will also provide novel data for the validation of mathematical models of the cervical spine. Presently models are somewhat limited by the scarcity of data – excepting kinematic data which is plentiful – for validation.\textsuperscript{11}

Although there have been many theoretical and experimental investigations of the biomechanics of cervical spine injury caused by direct impact to the head,\textsuperscript{13,16,35,36} there is very little data describing how compressive forces are transmitted through cervical joints and how applied loads are distributed between the anatomic elements.

Many groups have investigated load-sharing or isolated component loads in the lumbar spine. In this region, pressure sensors have been used to analyse intervertebral disc biomechanics\textsuperscript{9,14,15,27,32} and strain gauges have been used to measure loads at the facet joints, vertebral body and endplate.\textsuperscript{2,24,34} Theoretical models have also been focused on load-sharing in this region.\textsuperscript{10,34} In contrast, in the cervical spine, there is almost no basic experimental data on cervical spine disc pressure behaviour.\textsuperscript{7,23} There are also very few studies, that have focused on load-sharing.\textsuperscript{6,17,21}

Pal and Sherk\textsuperscript{17} have measured the load passing through the posterior and anterior columns of the cervical spine under compressive force application. They mounted load cells sequentially under the vertebral body anteriorly and under each facet joint posteriorly. They concluded that the posterior column was most highly loaded. Presumably to prevent buckling of their specimens only very low forces of approximately 26 N were applied. It is unclear whether their conclusions would be valid at the higher load levels present physiologically. Goel and Clausen\textsuperscript{6} recently performed a theoretical load-sharing analysis of the C5-6 motion segment using the finite element method. In contrast to Pal and Sherk,\textsuperscript{17} they found that the anterior column transmitted most of the applied compressive load (88%). Their approach allowed prediction of many important parameters such as disc pressure, force transmitted through the facet joints and
ligament strains. Very few experimental data were available to validate their model. These authors and others active in theoretical modelling of the cervical spine have noted that the lack of data about the loading of specific anatomic structures (i.e. disc, ligaments) limits the validity of the models.

Pintar et al. have conducted an experimental study of load-sharing in the cervical spine based on surface strains on the vertebral body and posterior elements. They applied pure and eccentric compressive loads to cervical spine segments and measured the resulting strains with uniaxial strain gauges (aligned with the cranial-caudal axis of the spine). This study represents a significant advance in the understanding of compressive force transfer through the cervical spine. However, no attempt was made to directly examine the behaviour of the intervertebral disc. Additionally, the use of uniaxial strain gauges implies that the strains of interest all occur cranial-caudally.

The objectives of our study were to analyse the load-sharing between the anterior column, represented by the vertebral bodies and the intervertebral disc, and the posterior column, represented by the facet joints and neural arch. This involved the use of experimental techniques to measure cervical disc pressure, vertebral body and facet joint strains. Our test protocol focused on the transmission of compressive force and its variation with changes in flexion-extension posture. A priority was to develop a loading protocol that applied compressive force at each posture but minimised the associated bending moment.

3.3. Materials and Methods

3.3.1. Anatomic Material

Six cervical functional spinal units (FSUs) were harvested, fresh-frozen and prepared according to established procedures. Two C2-3, two C4-5, one C3-4 and
one C5-6 FSU resulted. Not all donor information was provided. Donor 2 (one C2-3 and one C4-5 FSU) was an 80 year old woman who died of ovarian cancer. Donor 3 (one C2-3 and one C4-5 FSU) was an 83 year old man who died of cancer. None of the donors had suffered trauma involving the head or cervical spine nor had they suffered diseases associated with changes in bone quality. X-rays were used to screen for severe degeneration or other bony abnormality. The dens was removed from the C2 specimens to simplify moulding. The extreme superior and inferior portions of the FSUs were moulded in polymethylmethacrylate (PMMA) taking care not to contact the intervertebral disc (IVD), facet joints or capsule with the PMMA. A method was developed using optical guides and anatomical landmarks to achieve a repeatable moulding such that the specimens natural lordotic posture was maintained. Water was frequently applied to the exposed specimen surfaces to prevent the anatomic structures from dehydrating.\textsuperscript{20} Small absorbent swabs were used to ensure no water came into contact with the strain gauges.

3.3.2. Instrumentation for Strain and Disc Pressure Measurement

Load transmission paths were identified by instrumenting each FSU as illustrated in Figure 3-1. Tri-axial electrical resistance strain gauge rosettes (FRA-1-11, Tokyo Kenkyujo Co. Ltd., Tokyo, Japan) were glued to the anterior surface of the caudal vertebral body (rosette 3) and beneath the left and right facet joints on the lateral masses (rosettes 1 and 2 respectively). The gauges were arranged in a rectangular configuration (one gauge at 0°, one at 45°, and one at 90°) and mounted on an epoxy backing of diameter 4.5 mm. Each gauge had a nominal resistance of 120 Ω, a gauge length of 1.0 mm, a gauge factor of approximately 2.1, and a maximum extensibility of 3%. Small areas were prepared for application of the strain rosettes using a scalpel to remove soft-tissue and an absorbent swab to remove any fluid. Cyanoacrylate-based
glue (Histoacryl Braun, Melsungen, Germany) was used to bond the strain rosettes to the bone. The rosette-bone bond was checked throughout the test protocol using two techniques. The rosette signal became impossible to balance if the rosette-bone bond failed. The rosette signals were balanced before each test to check for this behaviour. In the second technique, a gentle pressure was applied to the bone adjacent to the rosette resulting in a small strain signal which would decrease to zero at a properly bonded rosette when the load was removed. This procedure was performed periodically throughout the experiment.

Owing to the small disc space in the cervical spine the measurement of intradiscal pressure in a manner that does not influence the biomechanics of the specimen is technically challenging. A novel method was developed to accomplish this and is described in detail in Chapter 2. To summarise, a disc-shaped miniature pressure transducer (Model 060 S, Precision Measurement Company, MI, USA) of diameter 1.5 mm and thickness 0.3 mm was inserted into the centre of the intervertebral disc through a 2.2 mm diameter needle. The needle was subsequently removed leaving only the lead wire passing through the disc annulus.

The pressure sensor and vertebral body strain rosette were considered to provide an indication of the force being transmitted through the anterior column of the spine while the two lateral mass rosettes did the same for the posterior column. Principal strain magnitudes and their orientation were calculated for each rosette. Disc pressure and bone strain data were collected at 10 Hz.

3.3.3. LOAD APPLICATION AND TEST PROTOCOL

Compressive force was applied using a bi-axial servohydraulic materials testing machine (Axial-torsional Mini-Bionix, MTS Systems Corp., Minneapolis, USA). A compression platen connected to the testing machine actuator applied compression to the specimen through a loading jig attached to the superior vertebra (Figure 3-2). The
loading jig also incorporated an indexed angular joint, which allowed the flexion-extension posture to be prescribed. The weight of the compression jig and upper moulding material (approximately 1.8 kg) was counterweighted to prevent it having any effect on the specimen's behaviour at low levels of applied compression.

Our loading apparatus was intended to subject the specimen to primarily axial compression and to superimpose a moment in the sagittal plane sufficient but not in excess of that necessary to maintain the flexion or extension posture. All other moments or forces were minimised. The loading jig allowed anterior-posterior translation of the specimen using four low-friction rollers. The spacing of the rollers was such that the four rollers were spaced considerably beyond the margins of the specimen. This allowed the force applied to each roller to equal that necessary for static equilibrium at the posture tested. The requirement of equilibrium ensured that the magnitude of the flexion or extension moment which was superimposed on the applied axial compression did not exceed that necessary to maintain the posture.

Prior to the main experiment, the specimens' full flexion and extension deflections under applied flexion and extension moments of 1.0 Nm were measured. The standard flexibility test protocol and pure, unconstrained moment application applied are described in detail in Chapter 4. The 1.0 Nm moment magnitude represented a compromise between the lowest (0.3 Nm) and highest (4.5 Nm) moments previously applied in vitro. In order to minimise the chance of specimen injury, moments in excess of 1.0 Nm were not applied. The compression test was performed at full flexion, neutral and full extension postures and normally at two degree increments between these extremes. In cases where the specimen had a small (i.e. less than 9°) flexion-extension range of motion (ROM) the test was performed at 1° increments.
At each posture, a compressive force of 800 N was applied at a rate of 10 N/s, held constant for 30 s and then removed at 10 N/s. Two preconditioning cycles were applied to allow viscoelastic effects to dissipate and data was collected from a third cycle. To evaluate the general repeatability and effects of test sequence on the measured quantities (whether the sequence in which the postures were tested affected the measured properties), the specimens were re-tested between one and five times per specimen at various postures and at various times within the testing sequence.

3.3.4. REACTION LOADS

To confirm that the moments and forces applied to the specimen conformed to that desired, the reaction loads were measured using a six-axis load cell (MC3A-6-1000, Advanced Mechanical Technology Inc., Watertown, MA, USA) mounted inferior to the specimen (Figure 3-2). According to the manufacturers specifications, the load cell was accurate to within ±12.5 N in the cranial-caudal direction and ±3.0 N in the horizontal plane and all moments were accurate to within ±0.07 Nm. Cross-talk was specified to be less than 2% of the channel reading. Because of restrictions on the number of analog signals we could simultaneously sample, the reaction moment about the specimen's cranial-caudal axis (y-axis in Figure 3-3) was measured using the MTS torque transducer instead of the six-axis load cell. The MTS transducer was accurate to within ±0.05 Nm. Isolated and combined moments and forces were applied to the load cells to verify the quoted specifications including accuracy and cross-talk.

The reaction moments (M_y) about the specimen's medial-lateral axis (x axis, Figure 3-3) were reported as changes with respect to the neutral posture for each specimen. This was done to counteract the effect of any initial anterior or posterior offset of the specimen with respect to the load cell's internal co-ordinate system. Reaction load data were collected throughout the load cycle at a frequency of 10 Hz.
3.3.5. Kinematics

Motion of the superior vertebra with respect to the inferior vertebra was measured using an infrared-sensitive optoelectronic motion analysis system (Optotrak 3020, Northern Digital Inc., Waterloo, Canada). Four infrared light emitting diodes (IRLED) were attached to the superior and inferior PMMA blocks (Figure 3-2). Each of the markers could be located in three-dimensional space with an accuracy of at least 0.15 mm. A lateral calibration x-ray was taken to allow anatomic co-ordinate systems to be identified at the middle of each endplate (Figure 3-3). The relative locations of the end-plates and IRLEDs were determined by digitising the relevant points using a backlighted digitising tablet (accuracy 0.2 mm, Digikon 24/36, Kontron Electronic AG, Zürich).

The locations of the IRLEDs were recorded throughout the measurement cycle at a frequency of 10 Hz. For each frame of data the location of the superior vertebra’s co-ordinate system with respect to the inferior vertebra’s co-ordinate system was calculated in terms of a three-dimensional translation vector and three ordered Euler angles using the ZYX ordering convention.\textsuperscript{26,29} Of particular interest was the variation in disc height (Figure 3-3).

3.3.6. Data Reduction and Statistical Analysis

The principal strains at each rosette where examined at each posture to determine if the strain field was dominated by tensile, compressive or shear strain. The quantitative criteria that were used to grade the strain states were:

a) If the tensile principal strain was at least 25% greater than the compressive principal strain and oriented within 30° of the cranial caudal axis the rosette was classified as dominated by cranial-caudal tension,
b) if the compressive principal strain was at least 25% greater than the tensile principal strain and was oriented within 30° of the cranial-caudal axis the rosette was classified as dominated by cranial-caudal compression and

c) if the principal strains were within 25% of each other and were oriented not closer than 15° from either the cranial-caudal or medial-lateral axis of the spine (i.e. within ±30° of the "pure" shear strain plane which is oriented 45° from both the cranial-caudal and medial-lateral axes) the rosette was dominated by cranial-caudal shear strain.

The smaller of the two principal strains was often considerably larger than what would have been expected based on a pure Poisson's contraction or expansion (i.e. approximately 30%). This is why the relatively conservative limits delineated above were placed on the conditions under which a rosette would be judged dominated by shear strain. Because the strain magnitudes exhibited a great deal of inter-specimen variation, presumably due to differences in bone quality and local vertebral geometry, the strain values for each specimen were normalised to the value measured at the neutral posture.1

Pearson product moment correlations were used to evaluate the strength of associations between variables. Where there was a clear rationale for selecting one variable to be independent, linear regression analysis was performed.4 Associations were considered significant at P<0.05.

3.4. Results

3.4.1. Bone strain

In general, bone strains increased linearly as the compressive load was applied. The strain response presented in Figure 3-4 was typical for a rosette dominated by compressive strain. The compressive principal strain, ε2, was aligned with the
specimen's cranial-caudal axis and was considerably greater (approximately 70%) than the tensile principal strain. The vertebral body rosettes were dominated by cranial-caudal compressive strain at all postures tested with the exception of one observation at 10° of extension at which compressive strain dominated but in a direction other than cranial-caudal.

The vertebral body rosettes exhibited an increase in compressive strain with increasing flexion posture and a decrease in compressive strain with increasing extension angles (Figure 3-5). This trend was significant across all postures \((r^2=0.54, P<0.0002)\) and exhibited a strong correlation between -5° extension and 4° flexion \((r^2=0.96, P<10^{-10})\). The average strain at the neutral posture \((-1414 \mu m/m\) was higher at this rosette than at the other rosettes.

In contrast to the vertebral body, the lateral mass rosettes were dominated by tensile, compressive, shear and by 'other' strains in almost equal proportion (Figure 3-6). Strains classified other did not satisfy any of the three criteria and were usually associated with low strain magnitudes. In 77% of the cases classified other, the principal strain magnitude was less than 200 \(\mu m/m\).

It was difficult to discern a consistent pattern among the lateral mass results but it was notable that only one rosette was dominated by tension while several were dominated by compression in the most extreme extension postures (-5° to -10°, Figure 3-6). In the most extreme flexion postures (7° to 14°), several rosettes were dominated by tension, shear or other strains but no rosettes were exposed to cranial-caudal compression (Figure 3-6). Individual specimen lateral mass results were usually more consistent. Many rosettes were dominated by tension or by compression throughout their ROM (Appendix 2).
The normalised strains for all lateral mass rosettes dominated by compression are plotted together in Figure 3-7. There was a significant increase in compressive strain with increasing extension angle ($r^2=0.62$, $P<0.04$). The average strain at the neutral posture (-186 µm/m) was lower for this case than for either the tension-dominated lateral mass rosettes (Figure 3-8) or the vertebral body rosettes (Figure 3-5).

At lateral masses dominated by tension (Figure 3-8), the tensile strain increased with increases in flexion angle ($r^2=0.91$, $P<0.0001$). The average strain at the neutral posture (323 µm/m) was higher for this case than for the compression-dominated lateral mass rosettes but lower than the vertebral body rosettes. Detailed principal strain orientation data is presented in Appendix 2.

At lateral masses dominated by shear, there was a weak but significant trend ($r^2=0.42$, $P<0.011$) of increasing shear strain as the extension angle increased and decreasing shear strain as the flexion angle increased (Appendix 2). The absolute average strain in the neutral posture was 222 µm/m and the maximum relative change with posture was a relatively modest 80% increase at 10° of extension and an 80% decrease at 5° of flexion.

A total of 36 observations were available for evaluation of the bone strain repeatability. The mean absolute difference in principal strain magnitude was 136.7 µm/m with a standard deviation of 188.9 µm/m. This represented 19% of the absolute strain magnitudes used in the comparison. There was no strong correlation ($r^2=0.11$, $P<0.02$) between the strain repeatability and the test sequence (i.e. number of tests between the two measurements used to evaluate repeatability). The orientation of the principal strains was highly repeatable with a mean difference of 8.4° and a standard deviation of 15.1°. There was no correlation ($r^2=0.02$, $P<0.36$) between the angle repeatability and the test sequence. Further bone strain data is presented in Appendix 2.
3.4.2. Disc Pressure

Due to technical problems, disc pressure was collected for only four of the six specimens. The disc pressure increased linearly with applied force. Any change in disc pressure with posture was too subtle to be resolved with the present protocol and number of specimens. The mean pressures at full extension, full flexion and neutral postures were within 0.35 MPa of each other (Figure 3-9). In the neutral posture repeatability tests, the mean pressure difference was -0.42 MPa with a standard deviation of 0.31 MPa. The normalised pressure signal was significantly correlated to test sequence (pressure normalised to the initial neutral posture value, $r^2=0.49$, $P<0.02$). The dependence on sequence made it impossible to determine if changes in pressure were due to sequence or posture. Further data regarding this phenomena is presented in Appendix 2.

3.4.3. Reaction Loads

At all postures, the average reaction forces were less than 16 N in the horizontal plane (z-x plane of the specimen). The average reaction moments are plotted as a function of posture in Figure 3-10. The reaction moments about the anterior/posterior axis (z-axis) were typically less than 2 Nm and had a maximum of 4.2 Nm. About the cranial-caudal axis (y-axis) the reaction moments were usually less than 1 Nm. The reaction moments about the medial-lateral axis (x-axis) exhibited different behaviours under flexion and extension postures. With increasing flexion angle the reaction moment was consistently small (< 1.1 Nm). The reaction moment increased with increasing extension up to a maximum of -10 Nm (extension moment) at 10° of extension. Reaction force data is presented in Appendix 2.
3.4.4. Kinematics

The testing apparatus prescribed the specimen's posture, in terms of its angular orientation, and the opto-electronic results confirmed that the specimen's posture did not change as the test progressed. Only the disc height (displacement along the cranial-caudal axis, Figure 3-3) changed as the compressive force was applied. The minimum disc height was measured at the maximum applied compression (800 N). In the neutral posture, the average disc height compression was 0.83 mm with a standard deviation of 0.44 mm. This corresponded to a compressive stiffness of 964 N/mm. At the neutral posture, test sequence was significantly correlated to the compressed disc height ($r^2=0.56$, $P<0.02$). The average decrease in the minimum disc height between the initial and final test in the neutral posture was -17.4% (Appendix 2).

3.5. Discussion

The objectives of this study were to identify the load-sharing paths in the cervical spine under compressive loading. Related objectives were to evaluate the influence of changes in flexion-extension posture on the load-sharing mechanisms and to minimise extraneous moments applied secondary to compressive force application.

The compression-dominated lateral mass rosettes exhibited no change in strain with increasing flexion but a clear increase with increasing extension was measured (Figure 3-7). The vertebral body strain decreased for extension postures (Figure 3-5) and the shear strain increased with increasing extension postures. Taken together, this behaviour indicated that the posterior column transmitted increasing amounts of compressive force as extension increased. The tensile-dominated lateral mass strains (Figure 3-8) and the compressive vertebral body strains (Figure 3-5) both increased as flexion posture increased. A mechanism of increasing anterior column load-sharing with increasing flexion is suggested by these results.
Buttermann *et al.* have also measured tensile strains at spinal facet joints. They determined that these strains were partially due to load transmitted through the facet joint capsule. It is likely that the tensile strains at our lateral masses also arose as a result of tension in the facet capsules. It is unclear why shear strains were also common at the lateral masses (Figure 3-6). Elementary mechanics suggests that these strains could result from laterally directed forces at the articular surface of the facet joints. Such forces could presumably result from friction at the facet joints coupled with lateral motion during compression or as a result of lateral deformation of the joint capsule. Further work is necessary to establish the physical mechanisms behind this strain state.

Although it was not possible to directly calculate the amount of force transmitted through individual components, several factors suggest that the vertebral bodies and intervertebral disc, representing the anterior column of the spine, usually transmitted the bulk of the applied compression. The strain states at the lateral masses varied considerably. Tensile strains and strains of negligible magnitude were measured even at some extension postures. This indicated that some lateral masses did not transmit compressive force at any posture. In contrast, the consistent and relatively large strains measured at the vertebral body rosettes suggest a consistent transmission of compressive forces. This conclusion is in agreement with the theoretical model of Goel and Clausen who estimated that 88% of applied compression loads pass through the intervertebral disc. The fact that intra-discal pressure did not appear to vary with changes in posture suggests that the compressive transmission through it was relatively constant. It may seem that the relatively invariant disc pressure contradicts the clear decrease in vertebral body strain with decreasing flexion and with increasing extension. We interpret this as a sign that the disc pressure indicates the force flowing through the whole vertebral body (i.e. the anterior spinal column). The drop in strain at the vertebral body's anterior surface may be matched by a corresponding increase at the posterior
aspects of the vertebral body. We did not measure bone strain at this location and so cannot quantitatively confirm this.

The reaction moment data indicated that an average extension moment of 10 Nm was necessary for equilibrium at a 10° extended posture. This was with respect to a coordinate system at the specimen's neutral-posture balance point (approximately at the geometric centre of the disc). The moment was potentially produced by two distinct mechanisms. In the first, additional moment may have been necessary for equilibrium due to stiffening of the intervertebral disc or other structures, which resist extension moment. This seems unlikely since no corresponding stiffening was detected under flexion loading. In the second possible mechanism, a posterior shift of the resultant compressive force vector from the neutral posture balance point would result in a posterior moment arm with respect to the neutral posture balance point. A 10 Nm moment corresponds to a 12.5 mm posterior shift of the resultant force. This is larger than half the endplate width in the anterior-posterior direction. This suggests that the vector may be posterior to the vertebral body at this posture and likely resulted in increased force transmission through the lateral masses. This is consistent with the increase in lateral mass compressive and shear strains measured at high extension angles.

Pintar et al. also applied 800 N of compression. In contrast to the results presented here they found compressive strains at the anterior vertebral body under anterior-eccentric compressive forces and tensile strains under posterior-eccentric loading. The tensile strains probably resulted from differences between the loading protocols. At their greatest posterior eccentricity, Pintar et al. applied compression at a point 3 cm posterior to the specimen's "balance point" (anterior-posterior location at which a point load in the sagittal plane produces no specimen flexion or extension). This resulted in a maximum extension moment of approximately 24 Nm, 14 Nm in
excess of the 10 Nm reaction extension moment which we measured at a 10° extension angle. It seems likely that the anterior tensile strains Pintar et al. measured were associated with the applied extension moment and that they were of sufficiently large magnitude to dominate the strain field and "overwhelm" the compressive strains due to the superimposed compressive force.

Because Pintar et al. applied force through smaller eccentricities in flexion, the resulting flexion posture and thus the loads applied in both protocols were similar. Our results at the vertebral body for this case are in close agreement with those of Pintar et al. for the corresponding posture (approximately 2000 μm/m at 10° of flexion in both studies).

Although they tested similar flexion postures, Pintar et al.\textsuperscript{21} reported very low (~100 μm/m) lateral mass strains for their most extreme flexion postures. They did not report tension strains. This inconsistency with our results was probably a result of differences in the data reduction techniques. Pintar et al. averaged the left and right lateral mass values and then averaged the resulting strains again for each posture. We feel that the technique used here, of analysing each lateral mass individually and normalising strain values allowed a somewhat more sensitive characterisation of the lateral mass strain states. Differences due to the use of uniaxial strain gauges in the Pintar study compared to triaxial gauges in the present study may also have contributed.

The bone strain results presented considerable challenges for interpretation. Strain magnitude is proportional to force applied, meaning highly strained rosettes indicated high load transfer. However, due presumably to inter- and intra-specimen variation in bone quality and anatomy, the absolute strain magnitudes exhibited considerable variation from specimen to specimen and between gauges mounted on the same vertebra. This was especially true at the lateral masses. Anatomical differences
also complicated our attempts to apply rosettes consistently in the same location. Therefore, it was not appropriate to directly compare absolute strain magnitudes to estimate which components were most highly loaded.

We applied two techniques to simplify the analysis and allow inter-specimen data to be combined. Strain values for each rosette, at each posture, were normalised to the corresponding value at the neutral posture. Our rationale for normalising to this value was that there were typically non-zero principal strain values at this posture. It was also expected that distinctly different strain behaviours would be evident in flexion and extension postures. The neutral posture provided a repeatable reference strain at an intermediate posture. Normalisation of vertebral bone strain values has been applied previously for similar reasons. The second technique was the classification of rosettes as compression-, tension-, or shear-dominated which used both principal strain magnitude and orientation information. This revealed that the lateral masses were exposed to a variety of distinct and posture-dependant loading modes. It would not have been possible to identify these loading modes with uni-axial strain gauges.

In contrast to the lateral masses, the vertebral bodies were consistently exposed to cranial-caudally-oriented compressive strains. At this location, the compressive strain magnitude varied linearly with changing posture between full flexion and extension postures. One potential reason for the excellent fit and small standard deviations between -5° extension and 4° flexion may have been that there were typically more data points for each posture in this region. This follows from the fact that all specimens had at least an 8° ROM (-4° to +4°) but fewer specimens could flex and extend to the more extreme postures. There were therefore fewer data points for these postures.

The principal strain magnitude and orientation were repeatable within an acceptable margin. The average error of 136 μm/m represents approximately 21% of the
maximum average strain observed at the tensile-dominated lateral mass gauges and 5% of the maximum average strain observed at the vertebral body rosettes. Thus the higher measured strain magnitudes were associated with a low error. It is important to note that the lower bone strain magnitudes in the study approach the limitations of the measurement method. Other investigators have reported similar variability when measuring vertebral bone strains using strain gauges.\textsuperscript{1,21}

The interpretation of disc pressure data as an indication of anterior column load-sharing was confounded by several factors. The correlation of disc pressure with test sequence implied that it was difficult to differentiate sequence-associated changes from posture-associated changes. Of course large variations in pressure (much greater than the variation due to sequence) would have been detected but our results suggest variations due to changes in posture were subtle if they existed at all.

The sequence dependence observed for normalised disc pressure and disc height suggest a viscoelastic alteration of the disc with a time constant in excess of the time for the complete test series. This occurred despite the two preconditioning cycles at each posture, the relatively slow load and unload rates and the 30 second hold at 800 N of applied compression. All of these steps were taken to minimise viscoelastic effects but they were insufficient. The lack of sequence dependence for the strain results suggests that the long-term viscoelastic changes to the disc did not affect the bone strain behaviour. Despite the sequence dependence, the average neutral posture disc stiffness was in close agreement with that measured by Shea \textit{et al.}\textsuperscript{25} (964 N/mm in this study compared to 957 N/mm).

In conclusion, our results suggest that the anterior column of the cervical spine transmits the bulk of the applied compressive forces at all flexion and extension postures of a specimen's physiologic ROM. The lateral masses become unloaded or are subjected to tension secondary to tension in the facet capsules in flexion postures, and
transmit increasing compressive force as extension increases. The force flow through the lateral masses increases at large extension angles but our results suggest that the bulk of the applied force continues to flow through the anterior column. The extension-associated compressive force transmission through the lateral masses is consistent with contemporary theories of facet joint injury secondary to axial compression during whiplash-associated neck extension. More work is necessary to determine under which conditions the posterior column force transmission becomes injurious.

3.6. References


3.7. Figures

Figure 3-1: FSU instrumentation, rosettes were numbered 1-3 (1 not shown).
Figure 3-2: Compression apparatus. The compression platen is fixed to the testing machine actuator. As the compression platen moves downward it applies compression to the specimen through the low friction rollers and loading jig. The flexion or extension posture was adjustable using the screws shown and index marks on the joint. Specimen kinematics and reaction loads were measured using the IRLED markers and six-axis load cell respectively. The specimen shown is a plastic replica of C4-5 being tested in 8° of extension.
Figure 3-3: Definition of the cranial and caudal vertebra co-ordinate systems. The dotted lines join the anterior- and posterior-most aspects of the appropriate endplate as identified from a lateral calibration x-ray. The co-ordinate systems were located at the centre of each of the dotted lines and were oriented such that their y-z plane was parallel to the plane made by the IRLEDs and with their z axes parallel to the testing apparatus as shown. The co-ordinate system's origin were located in the sagittal plane. This was a known distance from the plane made by the IRLEDs.
Figure 3-4: Principal strains and their orientation at the vertebral body rosette (rosette 3) for a C3-4 specimen in 10° of flexion. $\varepsilon_1$ and $\varepsilon_2$ are principal strains, oriented 90° from one another, and $\phi$ is the angle from gauge a ($\varepsilon_a$) to the first principal strain, $\varepsilon_1$. The rosette orientation is shown in the insert. Compressive strains are negative and tensile strains are positive.
Figure 3-5: Average normalised strain plotted as a function of posture for the vertebral body rosettes (rosette 3), which were dominated by compressive strain. The error bars represent one standard deviation. Data points with no error bars were from a single specimen. The trend of increasing strain with increasing flexion posture was highly linear between 5° of extension and 4° of flexion.
Figure 3-6: Strain states at the lateral mass rosettes. The number of observations of each type of strain are indicated by the bar heights for each posture.
Figure 3-7: Average normalised strain plotted as a function of posture for the lateral mass rosettes (rosettes 1 and 2), which were dominated by cranial-caudal compressive strain. The error bars represent one standard deviation. Data points with no error bars were from a single specimen.
Figure 3-8: Average normalised strain plotted as a function of posture for the lateral mass rosettes (rosettes 1 and 2), which were dominated by cranial-caudal tensile strain. The error bars represent one standard deviation. Data points with no error bars were from a single specimen.
Figure 3-9: Average intradiscal pressure at full flexion, full extension and neutral postures. Error bars indicate one standard deviation.
Figure 3-10: Average reaction moments measured beneath the caudal vertebrae. Moments are defined in accordance with the vertebral co-ordinate system (i.e. negative M values correspond to extension moments, Figure 2-3). Error bars indicate one standard deviation. Data points with no error bars were from a single specimen.
4.1. Abstract

Quantitative characterisation of load transmission paths in the cervical spine may allow confirmation or refinement of cervical spine injury mechanisms or even of mechanical factors that may be related to joint disease development. Lower cervical functional spinal units were subjected to flexion, extension, bilateral lateral bending and bilateral torsion moments of 1.0 Nm in vitro. Each moment direction was tested with and without a 200 N axial compression. Load transmission paths were identified by applying strain gauge rosettes to the anterior surface of the caudal vertebral body and beneath the left and right facet joints on the caudal lateral masses. A miniature pressure transducer was used to measure intradiscal pressure. Motions of the cranial vertebra with respect to the caudal vertebra were measured using an optoelectronic motion analysis system and described using Euler angles and helical axes of motion. In general, all strains increased with increasing moment application. The clearest trends were evident under flexion-extension loading. In flexion, the anterior vertebral body strains were primarily compressive (-140 µm/m) and the lateral mass strains were primarily tensile (80 and 170 µm/m). In extension the opposite was true. Under lateral bending the lateral masses on the side toward which bending occurred were more highly strained.
Axial torsion resulted in equal strains at both lateral masses. The addition of axial compression caused a relatively large compressive strain to be induced at the vertebral body rosette (approximately -450 μm/m) and much smaller increases in strain (< 100 μm/m) at the lateral masses. Intervertebral pressure was equal for flexion and extension loading (approximately 0.1 MPa) and was twice the lateral bending pressure and three times the torsion pressure. The addition of axial compression resulted in a disc pressure between 0.5 and 0.6 MPa which was invariant or decreased with applied moment for all cases except flexion (a 0.1 MPa increase). The helical axes of motion and range of motion data indicated that axial torsion and lateral bending were strongly coupled. Except in flexion, the addition of axial preload significantly reduced the specimen range of motion. The tensile strains measured were likely induced as a result of tension in the facet joint capsules or disc annulus. Our results suggest applied compression forces are transmitted primarily through the anterior column of the spine and the facet joints of the posterior column act to guide and constrain the kinematics of the segment in response to applied moments.

4.2. Introduction

The biomechanics of the cervical spine have been studied for many reasons. Traumatic injury mechanisms that occur as a result of head impact\textsuperscript{12,15} or whiplash,\textsuperscript{8,19,44} the effect of disease\textsuperscript{5} and the efficacy of surgical procedures\textsuperscript{25,31} have all been investigated. As a result of this work, the kinematics of the cervical spine, in terms of range-of-motion (ROM) and associated coupled motion magnitudes are well characterised. Several techniques have been developed to estimate traumatic and non-traumatic global loading. There is however, a paucity of data describing how applied loads are shared amongst the anatomic components of the cervical spine. This information could help clarify basic mechanisms of traumatic injury (e.g. during
whiplash), or allow refinement of theories on degenerative joint disease aetiology. In the theoretical realm, basic load-sharing data would provide investigators with a data set that could be combined with the abundant kinematic data, for model validation.

The effect of component transection\textsuperscript{22} or injury\textsuperscript{29} on the resulting specimen kinematics has been investigated. This approach provides an indication of the importance of the injured component for joint stability. This is clearly important for clinical diagnosis and choice of treatment. Associated information about the components' role in supporting the applied load is also implicit in the change of kinematics. However, load-sharing information obtained in this way is qualitative rather than quantitative since transducers to directly measure component load have not been applied.

Load-sharing has been the focus of several studies in the lumbar spine.\textsuperscript{30,32,43} However, little data of this type is available for the cervical spine. Pal and Sherk\textsuperscript{17} and Pintar et al.\textsuperscript{27} have measured the load passing through the posterior and anterior columns of the cervical spine under compressive force application. Goel and Clausen\textsuperscript{4} have performed a load-sharing analysis of the C5-6 motion segment using a finite element model. Bending moments were applied to the model with and without a compressive preload. They were able to predict many important parameters such as disc pressure; force transmitted through the facet joints and ligament strains. Because of the scarcity of load-sharing data this model was only validated with kinematic data. To our knowledge, load-sharing mechanisms in the cervical spine under applied bending moments have not previously been experimentally investigated.

We believe that any investigation of load-sharing through vertebral segments should include measurement of force transmission through the intervertebral discs in addition to other anterior or posterior column load measurement. The disc data provides an indication of force transmission through the anterior column of the spine. The measurement of intervertebral pressure is a common method for investigating force
transmitted through intervertebral discs. However, very little cervical disc pressure data is available. The reason for this may be the technical challenge associated with implanting sensor in the small disc space of the cervical spine (Chapter 2).

Hattori et al. and Pospiech et al. have measured in vivo and in vitro pressures respectively in cervical spines. Hattori et al. focused on the effects of posture and degeneration. Pospiech et al. examined whole cervical spines under a flexibility protocol. They determined the effects of simulated muscle action and ventral fusion. The variation of disc pressure in vitro using a functional spinal unit and flexibility protocol has not yet been documented.

Applied loads result in motion of the vertebral joint. This motion and the resulting change in posture likely influence the load-sharing mechanisms. For this reason, we believe that specimen kinematics should be included in analyses of load-sharing mechanisms. In addition to the common Euler angle approach, we used the helical axis of motion (HAM) method to help us interpret the load-sharing information. This method of kinematic analysis describes the motion of a rigid body as rotation about and translation along a unique axis (the helical axis, also called the screw axis).

The HAM is a three-dimensional analogue to the planar centre of rotation and has been used to describe the kinematics of thoracic and lumbar spine segments. HAM analysis has also been applied at other anatomic joints. Although some authors have hypothesised helical axis parameters based on cervical spine morphology. There is a paucity of quantitative data about the HAM in the cervical spine.

By virtue of the detailed information provided about the motion and the intuitive interpretation allowed by HAM analysis we feel it has the most potential for demonstrating the relationships between specimen kinematics and load-sharing. In
addition, because of the lack of HAM information for the cervical spine, this data will also provide novel information about cervical spine kinematics in its own right.

The objectives of our study were to characterise the load-sharing behaviour of the cervical spine under non-constraining pure moment application. Finite helical axes of motion and Euler angle kinematic analyses were also used to interpret load-sharing. The influence of compressive force representative of in vivo neck musculature was also evaluated.

4.3. Materials and Methods

4.3.1. ANATOMIC MATERIAL

Seven lower cervical functional spinal units (FSUs) were harvested, fresh-frozen and prepared according to established procedures. Two C2-3, two C4-5, one C3-4, one C5-6 and one C6-7 FSUs resulted. With the exception of the C6-7 specimen (which was added for this study) this was the same anatomic material described in Chapter 3. The C6-7 FSU donor (donor 1) was an 82 year old woman, 149 cm tall, 49 kg in weight, who died of a lung embolism. The procedures used for specimen moulding and handling are also described in that chapter.

4.3.2. INSTRUMENTATION FOR STRAIN AND DISC PRESSURE MEASUREMENT

Load transmission paths were identified by instrumenting each FSU as illustrated in Figure 3-1. Tri-axial strain gauge rosettes were glued to the anterior surface of the caudal vertebral body and beneath the left and right facet joints. Details of the rosette application and measurement techniques are provided in Chapter 3.

Owing to the small disc space in the cervical spine, the measurement of intradiscal pressure in a manner that does not influence the biomechanics of the specimen is technically challenging. A novel method was developed to accomplish this and has been described in detail in Chapter 2.
The pressure sensor and vertebral body strain rosette were considered to provide an indication of the force being transmitted through the anterior column of the spine while the two lateral mass rosettes did the same for the posterior column. Disc pressure and strain signals were sampled at 10 Hz. Principal strain magnitudes and their orientation were calculated for each rosette. 

4.3.3. LOAD APPLICATION AND TEST PROTOCOL

The specimens were subjected to isolated moments of 1.0 Nm using a custom load application apparatus consisting of computer-controlled pneumatic cylinders which produced pure moments by applying equal and opposite forces to the rims of 100 mm diameter pulleys (Figure 4-1). The pneumatic cylinders and associated hardware were mounted on a series of linear and rotational bearings. The bearings allowed the spine specimen to move without constraint, and ensured that the applied moments remained constant. Flexion, extension and bilateral lateral bending and torsion moments were applied. Each moment was applied in four equal steps of 0.25 Nm. At each step, the moment was sustained for 30 seconds to allow viscoelastic effects to dissipate. For each loading condition two preconditioning load cycles were applied to the specimen after which a third cycle was applied and used for data collection.

Each test was performed first without and then with a 200 N axial preload, which was applied to simulate the effect of neck musculature. To minimise artefact moment production the preload was applied using a type III preload as described in Appendix 5. Due to the small disc heights in the cervical spine, the cranial and caudal preload wire guides were aligned with the corresponding moulding material (Figure 4-1). It was impractical to align the guides with the specimen’s endplates as we did in the lumbar spine (Appendix 5). The preload force was applied through the specimen’s balance point which is the anterior-posterior point of load application that resulted in a pure
compressive displacement without any associated rotation when the preload was applied.  

4.3.4. Kinematics

Motion of the superior vertebra with respect to the inferior vertebra was measured using an optoelectronic motion analysis system sensitive to infrared light (Optotrak 3020, Northern Digital Inc., Waterloo, Canada). Four infrared light emitting diodes (IRLED) were attached to the superior and inferior PMMA blocks (Figure 4-1). Each of the markers could be located in three-dimensional space with an accuracy of less than 0.15 mm. Anatomic co-ordinate systems were defined using a lateral calibration x-ray as described in Chapter 3 (Figure 3-3).

The locations of the IRLEDs was recorded throughout the measurement cycle at a frequency of 10 Hz. For each frame of data the location of the superior vertebra's co-ordinate system with respect to the inferior vertebra co-ordinate system was calculated in terms of a three-dimensional translation vector and three ordered Euler angles using the ZYX ordering convention. The HAM for the specimen movement which started at no applied moment and ended at 1.0 Nm of applied moment was calculated as described by Kinzel et al. The HAM orientation, and the points at which it penetrated each of the planes of the caudal vertebra co-ordinate system (Figure 3-3) were calculated for each moment application direction and preload combination. Kinematic results were also evaluated using the animation technique described in Appendix 1. The motion of the cranial vertebra and the HAM, with respect to the caudal vertebra, were animated and displayed graphically.

4.3.5. Data Reduction and Statistical Analysis

The maximum and minimum principal strains at 1.0 Nm of applied load were calculated for each specimen and then averaged across the seven specimens, for each loading scenario. Changes in these parameters, with respect to the initial situation
before the moment was applied, were also calculated. Our rationale for reporting the change in strain and disc pressure data is related to the preload results. Preload typically caused initial strains to be induced at some rosette locations. The analysis of the change in strain with respect to the initial strain allowed us to evaluate whether the preload modulated the load-sharing response to the applied moment. If the change in strain or pressure was the same with and without preload we concluded that preload did not modulate the load-sharing mechanism.

Statistical power analyses indicated that an impractically high number of specimens would be necessary to identify expected differences in strain or disc pressure quantities as statistically significant under the inter-specimen variance conditions. For example more than 16 specimens would have been necessary to detect differences in strain of 100 μm/m with P<0.05 under the variance conditions measured. They were therefore not subjected to a statistical analysis. Matched pairs t-tests were used to examine the variation in specimen kinematics as a function of preload application. Statistical tests were considered significant at P<0.05.

4.4. Results

4.4.1. Bone strain

Strain behaviour between 0 and 1 Nm: Bone strains increased with increasing moment application. An example of how the average principal strain varied as a function of flexion and extension moment application is presented in Figure 4-2. At the vertebral body gauge, the application of flexion moments resulted in a dominant compressive strain field that became aligned (within 5°) with the cranial-caudal axis of the spine. In contrast, under extension loading a dominant tensile strain field was present which also became oriented cranial-caudally (within 15°). The tensile strain magnitude was three times greater than the compressive strain magnitude.
The addition of 200 N of compressive preload resulted in an average compressive strain before any moment was applied. The subsequent application of flexion moments resulted in a 56% increase in the average compressive strain. Extension moments resulted in a 49% decrease in compressive strain. The compressive strain was oriented within 5° of the specimen's cranial-caudal axis at all moment levels. Under moments other than flexion or extension, the principal strain orientations exhibited considerable inter-specimen variation and no consistent trends with respect to any anatomic axes could be identified. Data of this type is provided for all loading modes in Appendix 3.

**Average strains at 1 Nm:** The average principal strains, measured at 1.0 Nm, are presented in Figure 4-3 for each of the moment direction and preload conditions. Flexion application resulted in dominant-compressive principal strains at the vertebral body and tensile strains at both lateral masses. Preload caused a 360% increase in average compressive strain at the vertebral body and more modest increases (29%-left, 93% right) in tensile strain at the lateral masses.

Under extension without preload, the vertebral body rosette measured a tensile-dominant strain field. The lateral mass rosettes were subjected to a strain field typical of shear (approximately equal magnitudes of tensile and compressive principal strain). Preload caused a dominant-compressive strain at the vertebral body rosette and modest increases in lateral mass strains.

Under torsion and lateral bending moments without preload the lateral mass and vertebral body rosettes measured dominant-tensile or shear-like strains. Preload caused increases in lateral mass tensile strains, in most cases with smaller or negligible increases in the corresponding compressive strains. At the vertebral body rosette, compressive strain was produced as a result of preload application combined with lateral bending and torsion moments.
Average change of strain at 1 Nm: The change in principal strain compared to the strain present before the moment was applied is presented in Figure 4-4. This information is a valuable supplement to the strain magnitudes presented in Figure 4-3 since it allows examination of the influence of the applied moment independent of pre-existing strains arising from preload application or other effects. At both preload conditions, flexion induced increases in tensile strains at the lateral masses and increases in compressive strain at the vertebral body.

Under extension loading, the lateral masses were exposed to approximately equal increases in compressive strain at both preload conditions. At the vertebral body, an increase in tensile strain occurred without preload and a decrease in compressive strain was recorded with preload.

Under lateral bending, the lateral mass ipsilateral to the applied moment (i.e. the right lateral mass for right lateral bending) experienced greater increases in strain at both preload conditions than did the contra-lateral rosettes.

Under torsion loading, the left and right lateral masses experienced moderate increases in shear-like strain in cases without preload. Preload resulted in smaller strain increases. There was no obvious relationship between the lateral mass strains and the direction of the torsional moment.

For a given preload and moment combination the vertebral body usually experienced larger magnitudes of principal strain (Figure 4-3) than the lateral masses. Larger changes in strain as moment was applied were also observed at the vertebral body (Figure 4-4).

4.4.2. Disc pressure

Without preload, the disc pressure was approximately equal for the flexion and extension, right and left lateral bending, and right and left torsion pairs (Figure 4-5A). The flexion-extension pressure was approximately double the lateral bending pressure.
and triple the torsion pressure. Preload caused large (between six and 20 times) increases in pressure. There was no consistent influence of load direction on disc pressure with preload.

Flexion with preload caused a small increase in pressure (Figure 4-5B). In all other loading directions, the application of preload resulted in no change in pressure or (under left lateral bending) a decrease in pressure.

The average change in disc pressure as a function of moment application is presented for all loading cases in Appendix 3.

4.4.3. Kinematics

The average main and coupled ROM for each testing direction is presented in Table 4-1. The average ROM for flexion was greater than that for extension both with and without applied preload. Preload significantly reduced the specimen ROM in all testing directions except for flexion. Lateral bending and axial torsion were strongly coupled under both preload conditions. The average coupled ROM ranged between 97 % (torsion with preload) and 53 % (lateral bending with preload) of the main motion ROM. Average kinematic data for each loading mode is presented in Appendix 3.

The variation in disc height has implications for the interpretation of the disc pressure results. Under flexion loading, the disc height decreased an average of 0.65 mm (standard deviation: 0.98 mm). Under extension, the disc height also decreased (average: 0.24 mm, standard deviation: 1.89 mm).

In four instances the calculated HAM orientation or location data were obvious "outliers" and were excluded from the HAM analysis. This occurred for two specimens under extension with preload, for one specimen under flexion with preload, and for one specimen under left lateral bending. In these cases the axis was displaced from the specimen by 50 to 100 mm. The involved specimens typically exhibited a coupled
translation during the motion. Irregular cranial-caudal translations of 1-3 mm were common during flexion-extension motions but not during other moment applications.

The HAM for flexion and extension rotations, was approximately parallel to the medial-lateral axis (Figure 4-6A) The helical axis penetrated the posterior aspect of the caudal vertebral body (in the mid-sagittal plane) within the cranial third of the vertebra (Figure 4-6B). Neither the axis orientation nor location appeared to be influenced by moment direction or preload application.

The HAMs for lateral bending were oriented parallel to the sagittal plane and at an angle of approximately 20° to the transverse plane (Figure 4-7A). They penetrated the middle of the cranial vertebral body within the caudal quarter of the vertebral body or the cranial portion of the intervertebral disc (Figure 4-7B). Preload tended to cause the HAM to be more horizontally oriented than the non-preload cases. Preload also caused the left and right lateral bending HAMs to move laterally compared to the non-preloaded cases. The HAM for left lateral bending moved approximately 5 mm to the right and the HAM for right lateral bending moved approximately 5 mm to the left. An example of the animation results obtained under lateral bending is presented in Appendix 1.

The HAMs for axial torsion were similar to those for lateral bending. They were also parallel to the sagittal plane and somewhat more vertically oriented than the lateral bending HAMs (45° compared to 20°, compare Figure 4-7A and Figure 4-8A). The torsional HAMs penetrated the anterior quarter of the cranial endplate of the caudal vertebra and were usually centrally located (Figure 4-8B). Neither preload nor torsional moment direction appeared to consistently influence the HAM location or orientation.

4.5. Discussion

The understanding of load-sharing in the cervical spine may help clarify injury mechanisms and joint disease aetiology in this region. Load-sharing mechanisms under
applied moments were measured. The influence of preload on the load-sharing mechanisms was also established.

4.5.1. Flexion and extension load-sharing

Our bone strain and disc pressure results suggest that compression in the anterior column and tension bilaterally in the posterior column of the spine support flexion moments (Figure 4-2, Figure 4-3 and Figure 4-5). This is consistent with our expectations based on beam theory in which compression is observed on the concave free surface and tension is observed at the convex free surface of beams exposed to pure moments. The tensile strains at the lateral masses are probably transmitted as a result of tension in the facet joint capsules. Buttermann et al.² have previously established that tensile bone strains can be produced secondary to capsular tension during flexion loading. In extension loading, tension was produced in the anterior column and compression was produced in the posterior column. The anterior tension was likely transmitted through the disc as tension in the anterior disc annulus. Preload prevented tensile strains from developing at the vertebral body rosettes under extension loading (Figure 4-2B and Figure 4-3C).

A large compressive load was transmitted through the anterior column under flexion and extension with preload (Figure 4-3C). This contrasts sharply with the almost constant load transmission through the posterior column at both preload conditions (Figure 4-3A and B). This suggests that most of the applied compressive force is transmitted through the anterior column irrespective of the flexion or extension moment applied. This is also consistent with the behaviour under compression loading (Chapter 3) and shear loading (Chapter 5).

The intervertebral disc pressure data suggests that the disc transmitted more axial force under flexion and extension loading than it did under the other moment types. It is somewhat surprising that the intervertebral disc pressure was similar for flexion and
extension moments. The extension helical axis locations suggest that the cranial vertebra rotated about a point close to the spinal canal (Figure 4-6B) which should have caused the end plates to move apart, resulting in a reduction of pressure. We hypothesise that the tension in the anterior annulus prevented this from happening and instead caused a caudal translation of the cranial vertebra leading to an increase in disc pressure. This hypothesis is supported by the kinematic data. Under both extension and flexion moments, an average decrease in disc height (and thus compression of the disc) was measured. The discrepancy between HAM location and disc height change may be explained by motion of the HAM during moment application (The HAMs presented in Figure 4-6 are averages for 0 to 1 Nm). Examination of the extension animation data (Appendix 1) confirmed that the average extension HAM was located at approximately the anterior annulus between 0 and 0.5 Nm of applied moment. Preload reduced the rotation magnitude and prevented the production of tension in the annulus and there was thus no change in disc pressure under extension with preload application.

4.5.2. LATERAL BENDING LOAD-SHARING

Our results suggest that the ipsilateral facet joint (i.e. the right facet joint for right lateral bending) transmits increasing amounts of force and the opposite joint transmits decreasing amounts of force as lateral bending moments were applied (Figure 4-4A and B). The lateral bending rotation and coupled axial torsion led to disc deformation, which was manifested as increasing disc pressure and vertebral body strain. We believe that by resisting this deformation the disc also supported a portion of the applied moment.

The resulting strains were not uniformly aligned with the cranial-caudal axis as they were under flexion and extension loading which had the effect of producing almost equal tensile and compressive principal strains. We believe that this strain distribution may arise from a net shear load or from a biaxial combination of loads. The highly coupled kinematics of the specimens under these loads caused complex relative motion
between the vertebrae. Thus shear forces could arise from facet or other bony contact in the horizontal plane. These motion patterns may also result in tension in the ligaments, joint capsules or disc annulus. Due to the coupling, the tension could also be oriented in the horizontal plane or otherwise obliquely to the cranial-caudal axis. Either of these effects, or a combination of both, could result in a shear-like strain distribution.

Preload caused the resulting ROM to decrease and thus the corresponding disc deformation was smaller resulting in smaller changes in disc pressure (Figure 4-5B) and vertebral body bone strain (Figure 4-4C). The large compressive strain at the vertebral body indicated that the majority of the applied preload was transmitted through the anterior column irrespective of lateral bending rotation.

4.5.3. AXIAL TORSION LOAD-SHARING

In a similar manner to lateral bending it appears that forces arising from axial torsion are transmitted through both anterior and posterior columns. In the anterior column, the disc deformed as a result of the specimen rotation and the strongly coupled lateral bending which elicited increases in vertebral body bone strain and intradiscal pressure. Although we expect the contralateral facet to be more highly loaded for a pure axial rotation, the lateral mass behaviour was approximately the same on both sides (Figure 4-3A and B and Figure 4-4A and B). We hypothesise that this was caused by the very strong lateral bending coupled rotation (Figure 4-8A, Table 4-1). When the vertebral body rotates about the axial rotation HAM, the main torsion rotation caused the contralateral facet joints to approach each other. In contrast, the coupled lateral rotation caused the facet joints to move away from each other. The lateral mass strain results suggest these effects result in the same strain magnitudes irrespective of whether left or right torsion was applied.

Similarly to lateral bending and flexion-extension, axial preload resulted in the majority of the compressive force being transmitted through the anterior column.
In flexion, extension and lateral bending, our experimental results were in overall agreement with the theoretical load-sharing results presented by Goel and Clausen. Under axial torsion, the model diverged from our results. They predicted that the contralateral facet would transmit 41% of the applied load and the ipsilateral facet would be unloaded and we measured approximately equal force transmission at the two locations.

4.5.4. Disc Pressure

Our results compared well with those from Pospiech et al. and Hattori et al. Our results were within the range of results that Pospiech et al. reported for similar loading conditions. They also reported that the largest intradiscal pressures were measured for flexion and extension loading. Our preload results also fell within the range of in vivo results reported by Hattori et al. They reported higher pressures for extension postures (910 kPa) than for flexion postures (590 kPa). This level of agreement confirms that the 200 N preload applied in the flexibility (Chapter 4) and shear (Chapter 5) experiments was representative of in vivo compression.

4.5.5. Kinematics

The ROM and coupling ROM data presented here is in good agreement with that reported by other investigators. The HAM locations also correspond well to the planar centre of rotation locations presented by other authors for in vivo flexion-extension. The orientation of the HAM for lateral bending and axial torsion is in close accordance with that hypothesised by Penning and Milne based on vertebral anatomy. The location of the HAMs under lateral bending with preload (right half of the disc for left lateral bending and vice versa) is in agreement with that previously reported for the thoracic and lumbar spine. It is not obvious why this occurred for preloaded lateral bending but not for lateral bending without preload. However, the slightly more
horizontal orientation of the preloaded helical axes suggests that the preload acts to reduce the lateral bending-axial torsion coupling.

Translations are known to render the HAM location parameters of dubious interpretative utility. For this reason, three HAM penetration points that were outliers were excluded from the analysis. The affected loading cycles exhibited translations coupled with the main rotation motion.

The effect of preload on cervical spine kinematics has not previously been conclusively established. We used custom guides to apply the preload that minimised the artefact moment produced as a result of preload application method (Appendix 5). The reduction of ROM which was elicited as a result of preload application for all loading directions except flexion suggest the existence of physiologic mechanisms which cause the segment to stiffen. This behaviour has also been noted in the lumbar spine. Possible mechanisms include facet contact and/or locking and disc stiffening in response to increased pressure.

4.5.6. LIMITATIONS

The large standard deviations that occurred for bone strain and disc pressure measurement limited our ability to establish measured differences as being statistically significant. This level of variation in these variables is nevertheless common for experimental studies of this type. Because of the difficulty of obtaining cadaver material this limitation is unavoidable for studies of this type.

Differences in anatomic component geometry between the three gauge locations rendered it impossible to draw quantitative conclusions about the precise load magnitudes in each spinal column. For example, because the vertebral body has a much larger diameter than the lateral masses the strain induced for a particular force is expected to be smaller at the vertebral body than at the lateral masses. Differences in bone density or bone structure between the locations would exacerbate this effect.
Anatomical variation also rendered the precise positioning of the rosettes with respect to the anatomy difficult to accomplish or measure. These same factors likely contributed to the large inter-specimen variance that we measured.

High variability was observed in principal strain directions under loading scenarios other than flexion and extension. This is not surprising given the complexity of the vertebral anatomy and number and orientations of the ligaments through which force could be transmitted.

4.5.7. CONCLUSIONS

Our results suggest that flexion moments are transmitted as compression in the anterior column and tension in the posterior column. The opposite is true for extension. Lateral bending induces internal forces in the disc and causes increased load transmission through the lateral mass ipsilateral to the applied moment. Torsion induces internal forces in the disc and equal load transmission through the left and right lateral masses. Compressive preload causes the ROM to decrease in all directions except flexion. Preload is transmitted primarily through the anterior column of the spine. The facet joints become loaded in response to applied moments and may act primarily to guide the specimen motion.

4.6. References


4.7. Figures

Figure 4-1: Moment application apparatus. The four IRLEDs mounted to the moulding material of each vertebra are visible. The anterior of the specimen is at left and its left side is visible. Steel cables were used to apply forces to the rims of the pulleys mounted above the specimen. The preload fixation points and guides were adjusted to be even with the surfaces of the moulding material.
Figure 4-2: Effect of moment application on the average principal strains and their orientation at the vertebral body rosette (rosette 3). $\varepsilon_1$ (the maximum principal strain) and $\varepsilon_2$ (the minimum principal strain) are oriented 90° from one another and $\phi$ is the angle from gauge a ($\varepsilon_a$) to the first principal strain, $\varepsilon_1$. The rosette orientation is shown in the inserts. Compressive strains are negative and tensile strains are positive. A) Flexion and extension moments B) Flexion and extension moments with preload.
Figure 4-3: Average principal strain magnitudes at 1.0 Nm of applied moment. The error bars indicate one standard deviation. PL=preload, NO PL=without preload, L. Bend=lateral bending. A) Rosette 1, Left lateral mass rosette. B) Rosette 2, Right lateral mass rosette. C) Rosette 3, Vertebral body rosette.
Figure 4-4: Average change in the principal strain magnitudes between 0 and 1.0 Nm of applied moment. The error bars indicate one standard deviation. PL=preload, NO PL=without preload, L. Bend=lateral bending. A) Rosette 1, left lateral mass rosette. B) Rosette 2, right lateral mass rosette. C) Rosette 3, vertebral body rosette.
Figure 4-5: A) Average intradiscal pressure at 1.0 Nm of applied moment. B) Average change in the intradiscal pressure between 0 and 1.0 Nm of applied moment. PL=preload, NO PL=without preload, L. Bend=lateral bending. Error bars indicate one standard deviation.
Figure 4-6: A) Orientation and B) Penetration location of the HAM for flexion and extension moments. A representative vertebra is rendered in the graphs. The HAMs shown describe motions which occurred between 0 and 1.0 Nm of applied moment. E=extension, F=flexion, EPL=extension with preload, FPL=flexion with preload. The error bars indicate one standard deviation.
Figure 4-7: A) Orientation and B) Penetration location of the HAM for lateral bending moments. Representative vertebrae are rendered in the graphs. The HAMs shown describe motions which occurred between 0 and 1.0 Nm of applied moment. L=left lateral bending, R=right lateral bending, LPL=left lateral bending with preload, RPL=right lateral bending with preload. The error bars indicate one standard deviation.
Figure 4-8: A) Orientation and B) Penetration location of the HAM for torsion moments. Representative vertebrae are rendered in the graphs. The HAMs shown describe motions which occurred between 0 and 1.0 Nm of applied moment. L=left torsion, R=right torsion, LPL=left torsion with preload, RPL=right torsion with preload. The error bars indicate one standard deviation.
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*Lateral bending and torsion coupling always resulted in rotations in the same direction as the main motion. Left axial torsion resulted in left lateral bending and vice versa and right axial torsion resulted in right lateral bending and vice versa.

Table 4-1: Average main and coupled ROMs. Pairs of superscripted letters indicate statistical significance at P<0.05.
CHAPTER 5. BIOMECHANICS OF THE HUMAN CERVICAL SPINE UNDER SHEAR LOADING:
INFLUENCE OF SUPERIMPOSED AXIAL COMPRESSION

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5.1. Abstract

By virtue of the relative slenderness of the neck and the weight of the head, the cervical spine can be subjected to considerable shear forces. Knowledge about the basic biomechanics and load-sharing of the cervical spine may help to elucidate injury mechanisms or even suggest mechanisms leading to cervical joint degeneration. The aim of this study was to characterise the load transmission paths through cervical functional spinal units (FSUs) as a function of anterior, posterior and lateral shear force application and to evaluate the effect of axial preload on specimen kinematics and load-sharing. Seven lower cervical functional spinal units were subjected to shear forces of 200 N. The force was applied to the cranial vertebra at an application point in the mid-disc plane. Each shear force was applied first without and then with a superimposed 200 N axial compression force. Load transmission paths were identified using strain gauge rosettes, mounted beneath each facet joint posteriorly and on the vertebral body anteriorly, and a miniature disc pressure sensor. Kinematics were measured using an optoelectronic motion analysis system. Anterior shear resulted in compressive strain at
the vertebral body, shear strain at the lateral masses and a small reduction in disc pressure. Posterior shear resulted in tensile force transmission at the vertebral body, shear or compressive force transmission at the lateral masses and an increase in disc pressure. Under lateral shear, all measured strains were consistent with shear force transmission and disc pressure increased. Preload resulted in an increase in compressive strain anteriorly and an increase in disc pressure. With preload, the lateral masses transmitted less force or unchanged magnitudes of force compared to the non-preloaded tests. The specimens had higher shear stiffnesses under posterior shear than under anterior shear. Under lateral and posterior shear, preload caused a decrease in specimen stiffness and an increase in shear displacement. Our results suggest that under anterior shear the cranial vertebra’s facet joints ramp up the caudal facet joints, the specimen flexes and pivots about the compressed disc annulus anteriorly and the disc experiences a cranial-caudal distraction. Posterior shear is transmitted as tension in the anterior annulus and may also be transmitted as tension in the anterior facet capsules. Under lateral shear, the cranial facet joint impinges against the caudal lamina and a large portion of the lateral shear force may be transmitted through this contact. Preload causes increased segment motion, decreased force transmission through the posterior column and increased compressive force transmission through the anterior column. Our results suggest the disco-ligamentous cervical spine may be rendered less stable and the posterior column may transmit less force with increasing neck muscle activation.

5.2. Introduction

By virtue of the relative slenderness of the neck and the weight of the head, the cervical spine can be subjected to considerable shear forces. This can result from direct impact to the head (from a fall, dive etc) or from non-impact “inertial” loading due to acceleration of the trunk in the absence of head support (as in whiplash loading).
Elementary mechanics dictates that this kind of loading appears as a shear force combined with a bending moment resulting from the moment arm between the applied force and the cervical level of interest. The surrounding neck musculature likely supports some of the applied loads but the shear component may still be large (i.e. between 100 and 2000 N). Knowledge about the basic biomechanics and load-sharing of the cervical spine may help to elucidate injury mechanisms or even suggest mechanisms leading to cervical joint degeneration. This type of data will also be a new source of information that is needed for the validation of contemporary mathematical models of the cervical spine.

Considerable effort has been expended to characterise the biomechanics of the lumbar spine under shear loading. Fewer investigations have focused on the biomechanics of the cervical spine under this loading mode. Panjabi et al. and Moroney et al. have characterised the kinematic response of cervical functional spinal units (FSUs, consisting of two adjacent vertebrae and the disc and ligament tissues that connect them) subjected to anteriorly-, posteriorly- and laterally- directed shear forces. Moroney et al., Panjabi et al. and Raynor and colleagues have measured the effect of ligament or posterior element injury on the resulting kinematics.

Although the previous work represents a baseline of knowledge regarding the shear kinematics in the cervical spine, there are also limitations and some basic issues that have not been investigated. Force must be applied to a motion segment in the mid-disc plane to result in pure shear forces. Other application points are known to subject the FSU to combined shear force and moment loading. Only Maroney et al. have applied shear forces to the cervical spine in the mid-disc plane.

Both Moroney et al. and Panjabi et al. applied relatively low magnitude shear forces (ca. 20 and 50 N respectively). These are much smaller than those expected in vivo for some physiologic (up to 135 N) or traumatic (1100-2000 N) loading scenarios.
Additionally, although small preloads have been included in some previous investigations\textsuperscript{16,20} the influence of axial preload representative of that present \textit{in vivo} (ca. 100-1000 N)\textsuperscript{15} has not been systematically investigated to our knowledge. Similarly, we could find no experimental data about load-sharing and intervertebral disc biomechanics as a function of applied shear forces in the cervical spine.

The objectives of our study were to experimentally characterise cervical spine load-sharing mechanisms, as a function of anterior, posterior and lateral shear force application. The influence of axial preload on specimen kinematics and load-sharing was also investigated.

\section*{5.3. Materials and Methods}

\subsection*{5.3.1. ANATOMIC MATERIAL AND LOAD-SHARING INSTRUMENTATION}

Details of the seven cervical functional spinal units (FSUs), and the associated screening and handling procedures are provided in Chapters 3 and 4. Load transmission paths were identified by using bone-mounted tri-axial strain gauge rosettes (Diameter 4.5 mm) and a miniature intradiscal pressure sensor as illustrated in Figure 3-1. The vertebral body rosette and the disc pressure sensor provided indications of anterior column load-sharing while the left and right lateral mass rosettes indicated posterior column load-sharing. Further details about the strain rosettes and the associated application technique are provided in Chapter 3.

Owing to the small disc space in the cervical spine the measurement of intradiscal pressure in a manner that does not influence the biomechanics of the specimen is technically challenging. A novel method was developed to accomplish this and has been described in detail in Chapter 3. To summarise, a disc-shaped miniature pressure of diameter 1.5 mm and thickness 0.3 mm was inserted into the centre of the intervertebral disc through a 2.2 mm diameter needle. The needle was subsequently
removed leaving only the lead wire passing through the disc annulus. Disc pressure and strain signals were sampled at 10 Hz. Principal strain magnitudes and their orientation were calculated for each rosette.\textsuperscript{8}

**5.3.2. LOAD APPLICATION AND TEST PROTOCOL**

The specimens were subjected to anterior, posterior, left and right shear forces using an adjustable loading jig (Figure 5-1) and a uniaxial servohydraulic materials testing machine (Model 358.10, MTS Systems Corp., Eden Prairie, MN, USA). We defined anterior shear as an anteriorly-directed force applied to the cranial vertebra. Posterior shear was the opposite. The shear force was produced using the testing machine actuator, which was attached to one end of a steel cable. The other end of the cable was attached to a load bracket screwed to the cranial vertebra's moulding material. The point of load application with respect to the FSU was continuously adjustable in the cranial-caudal direction using a linear bearing. The load application point was adjusted so that the shear force (F in Figure 5-1) was applied in the mid-disc plane. The applied shear force was measured using the testing machine load cell (Accuracy < 1.0 N, model 661.19 F03, MTS Systems Corp., Eden Prarie, MN, USA).

For each load cycle an initial force of 20 N was applied to provide a common initial force condition. Then the force was increased at a constant rate of 5 N/s to 200 N. The force was held constant for 30 s to allow viscoelastic effects to dissipate.\textsuperscript{72} Finally the force was released at 5 N/s. Two preconditioning load cycles were applied and a third cycle was used for data collection.\textsuperscript{17} The caudal vertebra was rigidly fixed and the cranial vertebra was unconstrained. The load bracket and associated hardware could be rotated about the specimen to allow the application of left, right, anterior or posterior shear. We defined anterior shear as an anteriorly directed force applied to the cranial vertebra.
Each direction of shear was tested without axial preload and then with a constant preload of 200 N. The preload was applied using a pneumatic cylinder which acted against the cranial vertebra through a roller. The roller allowed anterior-posterior (for anterior/posterior shear) or lateral (for left/right shear) translation of the cranial vertebra (Figure 5-1). The preload force was initially applied through the specimen's balance point, which is the anterior-posterior point of load application that resulted in a pure compressive displacement without any associated rotation when the preload was applied.27,31

Shear and preload force magnitudes were selected to be representative of those believed to occur in vivo during non-traumatic activities as a result of head weight and muscle activation.15 Although both higher and lower loads were predicted by Moroney et al.,15 we felt that the loads selected were high enough to produce measurable bone strain and disc pressure signals but not high enough to cause bone, soft-tissue or moulding failure.25

5.3.3. Kinematics

Motion of the superior vertebra with respect to the inferior vertebra was measured using an optoelectronic motion analysis system sensitive to infrared light (Optotrak 3020, Northern Digital Inc., Waterloo, Canada). Four infrared light emitting diodes (IRLED) were attached to the superior and inferior PMMA blocks (Figure 5-1). Anatomical co-ordinate systems were identified for each vertebra as described in Chapter 3 (Figure 3-3). The locations of the IRLEDs were recorded throughout the measurement cycle at a frequency of 10 Hz. For each frame of data the location of the cranial vertebra's co-ordinate system with respect to the caudal vertebra's co-ordinate system was calculated in terms of a three-dimensional translation vector and three ordered Euler angles using the ZYX ordering convention.26 Specimen kinematics were also examined using the animation procedure described in Appendix 1.
5.3.4. Data reduction and statistical analysis

Data at 5.0 N increments of applied shear force between 20 and 200 N was selected from the continuous kinematic, bone strain and disc pressure data for analysis. This allowed direct comparison of the various parameters at each applied load level. Shear stiffness was calculated for each specimen between 20 and 40 N and between 180 and 200 N of applied anterior and posterior shear. Under lateral shear, the shear stiffness was calculated between 100 and 120 N. These levels of force represented regions within which the force-displacement relationships were typically linear. Linear regression analyses were applied to identify the best fit to the force-displacement data.

Statistical power analyses indicated that an impractically high number of specimens would be necessary to identify expected differences in strain quantities as statistically significant under the inter-specimen variance conditions. For example, more than 16 specimens would have been necessary to detect differences in strain of 500 μm/m with P<0.05 under variance conditions typical of those measured (standard deviation = 500 μm/m). They were therefore not subjected to statistical analysis. Differences in kinematic or disc pressure quantities were tested for statistical significance using matched pairs t-tests. Appropriate correction for multiple comparisons was applied where necessary.7 Left and right kinematic and intervertebral shear data were tested for symmetry and combined when they were found to be symmetric.

5.4. Results

5.4.1. Bone Strain

For non-preloaded tests, the bone strains generally increased approximately linearly with increasing shear force application. Preload caused an initial bone strain, which typically also increased with increasing shear force application. The orientation of the principal strains exhibited considerable variation between and within loading cases,
rendering interpretation difficult. The average variation of principal strain and principal strain orientation as a function of shear force application is presented for each loading case in Appendix 4.

In anterior shear, the greatest average strains were measured at the left lateral mass location (Figure 5-2A). The maximum (tensile, $\varepsilon_1$) and minimum (compressive, $\varepsilon_2$) principal strains were approximately equal in magnitude, suggesting a shear-type strain distribution. The strains were smaller at the right lateral mass and indicative of a dominant compressive strain field ($\varepsilon_2=1.8\varepsilon_1$). At the vertebral body the strain field was highly compressive-dominant ($\varepsilon_2=3.5\varepsilon_1$). Preload caused a decrease in strain at the left lateral mass and no change in strain at the other rosette locations.

Under posterior shear, the greatest principal strains were measured at the right lateral mass gauge and were consistent with a shear strain distribution. The left lateral mass was exposed to a compressive-dominant strain field ($\varepsilon_2=2\varepsilon_1$). At the vertebral body tensile strain dominated (2.1$\varepsilon_2$). Preload caused a reduction in strains at the lateral masses as well as a shift towards tensile-dominated strain fields ($\varepsilon_1=3.2\varepsilon_2$, right and $\varepsilon_1=1.4\varepsilon_2$, left). At the vertebral body rosette, preload caused a compressive dominated strain ($\varepsilon_2=3.2\varepsilon_1$).

Under lateral shear, non-preloaded right shear produced the largest lateral mass strain. Lateral mass strain distributions were indicative of shear strain at all shear and preload conditions except for right shear where tensile strain was dominant ($\varepsilon_1=2.1\varepsilon_2$ without preload, $\varepsilon_1=1.7\varepsilon_2$ with preload). Under left and right shear at the vertebral body rosette, shear strains were present without preload. Preload caused the strain field to become compression dominated ($\varepsilon_2=1.7\varepsilon_1$ left, $\varepsilon_2=3.8\varepsilon_1$ right).

The average total change in bone strain in response to shear force application is plotted in Figure 5-3. Under non-preloaded shear the change in strain was approximately
the same as the absolute values presented in Figure 5-2. The change in strain for the preloaded cases was different from values presented in Figure 5-2 indicating that an initial preload-associated strain was present before the shear force was applied. At the lateral masses, the change in strain due to shear load application was always small (between 18 and 220 \( \mu \text{m/m} \)). At the vertebral body, a decrease in \( \varepsilon_2 \) was recorded under posterior shear with preload and an increase in \( \varepsilon_2 \) was recorded under anterior shear with preload. Under lateral shear, preload caused increases in both \( \varepsilon_1 \) and \( \varepsilon_2 \) (left shear) or a reduction in \( \varepsilon_2 \) (right shear).

5.4.2. INTRADISCAL PRESSURE

Non-preloaded anterior shear resulted in a small negative intradiscal pressure (Figure 5-4A). In contrast under posterior shear, a positive disc pressure was recorded. Preload caused positive increases in pressure under both anterior and posterior shear. The largest pressure was recorded for posterior shear with preload (420% greater than anterior shear with preload and 142% greater than non-preloaded posterior shear). The disc pressure was not significantly different under left and right shear and lateral shear data was therefore combined. Under lateral shear, positive disc pressures were measured and preload caused a 300% increase in disc pressure.

Examination of the change in disc pressure as a result of shear force application reveals that the disc pressure decreased under anterior shear for both preload conditions (Figure 5-4B). In contrast, posterior and lateral shear caused approximately equal (within 38% for posterior shear and 23% for lateral shear) increases in disc pressure for both preload conditions. The average disc pressure is plotted as a function of applied shear force for all loading cases in Appendix 4.
5.4.3. Kinematics

Non-preloaded anterior shear caused a coupled rotation about all three axes and a coupled cranial translation (Table 5-1). Anterior shear with preload and posterior shear under both preload conditions caused little coupled motion (less than 1° of rotation and 0.33 mm of translation). Lateral shear was associated with coupled lateral bending and axial torsion rotations (range 2.7° to 4.7°).

Non-preloaded anterior shear caused the greatest shear translation (>2.0 mm, Figure 5-5A). Preload caused a non-significant decrease in translation (18%). Posterior shear resulted in a 70% decrease in translation compared to anterior shear. Under both posterior and lateral shear, preload caused a significant increase in shear displacement (62% posterior, 51% lateral).

The force-displacement relationships were typically highly linear within the force ranges analysed (20-40 N and 180-200 N for anterior and posterior shear and 100-120 N for lateral shear). The average coefficients of determination ($r^2$) ranged from 0.97 (for anterior shear, 20-40 N) and 0.99 (for anterior shear with preload, 180-200 N). Average force-displacement and force-rotation results are presented in Appendix 4.3.

With the exception of applied shear forces less than 40 N, preload consistently resulted in a decrease in shear stiffness (Figure 5-5B). The preload-associated decrease was only significant under posterior shear (180-200 N, a 26% decrease) and lateral shear (a 37% decrease). Except for anterior shear with preload, anterior and posterior shear stiffnesses were higher at the larger shear force (180-200 N) than at the lower shear force (20-40 N). This effect was only significant for the non-preloaded condition (a 360% increase under anterior shear and a 97% increase under posterior shear).
Animation frames at an applied shear load of 200 N for a typical specimen are presented in Figure 5-6. Under anterior shear the facet joints were pressed into contact and the cranial vertebra was slightly flexed (Figure 5-6A). Under posterior shear the facet joints were distracted (Figure 5-6B). Under lateral shear an impingement of the cranial facet joint on the caudal lamina contralateral to the shear force application direction was identified (Figure 5-6C). The impingement and subsequent ramping up of the facet on the lamina was reflected as a local increase in stiffness on the force displacement plot (Appendix 4).

5.5. Discussion

This study identified the major load-sharing mechanisms of the cervical spine under shear loading and characterised the influence of axial preload on load-sharing and kinematics.

5.5.1. LOAD-SHARING

The animation technique enabled a graphical synthesis of the relationships between specimen anatomy, load-sharing parameters and specimen kinematics. Our results suggest that anterior shear force resulted in a "ramping" of the cranial facet joints against their caudal Figure 5-6A). The corresponding force transmitted through the facet joints was manifested as either compressive or shear forces at the lateral mass rosettes (Figure 5-2A and B). The ramping of the facet joints was associated with a flexion rotation of the cranial vertebra, which resulted in compression at the anterior vertebral body (Figure 5-2C). The cranial vertebra thus appears to "pivot" about the compressed disc annulus anteriorly (Figure 5-6A). The posterior ramping and anterior pivoting resulted in a distraction of the intervertebral disc manifested as an anterior translation of the cranial vertebra (Table 5-1) and a decrease in disc pressure (Figure 5-4B).
Disc pressure reduction in response to shear loading has not previously been reported to our knowledge. Studies in our laboratory\(^5\) and by other groups\(^3\) have established that disc pressure increases under anterior shear in the lumbar spine. The difference in disc pressure behaviour between regions may be explained by differences in facet geometry. The cervical facet joints make a smaller angle with the transverse plane allowing easier “ramping” of the cranial vertebra in response to anteriorly-directed forces.\(^{19}\) The posterior annulus and posterior longitudinal ligament may also be subjected to tension under anterior shear but we did not have sensors appropriately placed to detect this.

Posterior shear resulted in tension at the vertebral body rosette (Figure 5-2C) which was likely transmitted through tension in the disc annulus and the anterior longitudinal ligament (ALL). The compression measured at the lateral mass gauges (Figure 5-2A and B) is somewhat surprising since the facets were distracted under posterior shear loading (Figure 5-6B). We hypothesise that tension in the anterior facet capsule could subject the lateral mass to a sagittal-plane bending moment, which in turn, resulted in the compressive strain observed at the lateral masses (Figure 5-2A). Posterior shear resulted in an almost pure posterior translation (Table 5-1). The intervertebral disc was therefore subjected to a pure shear deformation, which was manifested as an increase in disc pressure (Figure 5-4B).

Under lateral shear, the applied force was transmitted primarily as shear forces at all rosette locations. The lack of a consistent direction-dependent increase in lateral mass strain (Figure 5-2A and B) suggests that the force transmitted through the facet-lamina impingement (Figure 5-6C) may not be detectable using the sensors we applied. The increase in disc pressure under this loading mode (Figure 5-4A) reflects the absence of any disc distraction or disc unloading mechanisms under lateral shear. Both
shear deformations of the disc and deformation associated with the coupled torsional and lateral bending rotations (Table 5-1) are reflected in the pressure increase.

Preload generally caused either a decrease or no change in strain at the lateral masses and an increase in disc pressure and compressive strain at the vertebral body. (Figure 5-2 and Figure 5-4) This suggests axial preload is preferentially transmitted through the anterior column and the facet joints may act principally to guide the segment motion. This is consistent with the results presented in Chapters 3 and 4. It is tempting to hypothesise the lateral masses transmitted less force when preload is applied because the specimen became stiffer in response to the preload (Appendix 5) and thus moved less. This is inconsistent with our results, however.

5.5.2. Kinematics

The increase in non-preloaded shear stiffness under anterior and posterior shear (Figure 5-5B) is typical of the non-linear increases in stiffness with increasing force exhibited by most biological soft tissues including spinal ligaments and the FSU itself. The stiffness behaviour in anterior shear between 20 and 40 N was distinct from the other load levels and shear force directions. The non-preloaded stiffness was much lower than for the other load cases and preload caused a 600% increase in stiffness compared to decreases for the other load cases. This suggests the existence of a mechanism which produces high initial (i.e. low force) resistance to anterior shear. Our results suggest that contact between the facet joints, resulting from a combination of shear and compression may be responsible for this behaviour. In the other loading modes the shear force does not directly force the facets into contact (Figure 5-6). The same facet joint interaction may be responsible for the reduction in shear displacement, which occurred as a result of preload application under anterior shear (Figure 5-5A).

The preload-associated decrease in stiffness for the posterior, lateral and anterior (180-200 N) loading cases (Figure 5-5B) are in accordance with the increases in
displacement for those cases (Figure 5-5A). This behaviour contradicts the preload-associated increases in shear stiffness measured in the lumbar spine\textsuperscript{11,28} and the increases in bending stiffness with preload observed under moment application in lumbar (Appendix 5) and cervical (Chapter 4) motion segments. In the lumbar spine, Panjabi et al.\textsuperscript{18} have also reported preload-associated decreases in stiffness under anterior and posterior shear. However, they applied the shear force cranial to the mid-disc plane resulting in a combined shear and flexion moment at the disc.

Yang and Begeman\textsuperscript{32} have proposed a mechanism for the preload-associated decrease in stiffness that we measured. They hypothesised that the intervertebral joints are compressed secondary to sudden straightening of the thoracic spine kyphosis. This attempts to accelerate the head cranially. The joint compression is thought to slacken the intervertebral ligaments leading to a decrease in segmental shear stiffness. Although it is not possible to identify the mechanism from our results, the decrease in stiffness, at the quasi-static loading rates we applied, is confirmed. Based on this, further examination of the preload-associated stiffness decrease to identify possible roles in traumatic injury or degeneration mechanisms is warranted.

Our results do not allow for a conclusive answer to the question of whether anterior or posterior shear provides the greatest overall stiffness. Under non-preloaded shear the shear stiffness was greater under posterior shear than it was under anterior shear (Figure 5-5B) and this was reflected in the significantly greater displacement under anterior compared to posterior shear (Figure 5-5A). This is a surprising result because most authors have hypothesised that the facet joints act as a physical "brake" to anterior displacement.\textsuperscript{11,28} The lumbar spine has been found to be stiffer under pure anterior shear than under posterior shear (both applied in the mid-disc plane).\textsuperscript{28} Facet joint orientations suggest a more effective physical block to anterior shear in the lumbar spine than they do in the cervical spine.\textsuperscript{19} In contrast to our results, Moroney et al.\textsuperscript{16} have
found greater cervical spine shear stiffness under anterior than under posterior shear but at significantly lower shear loads than were applied in this study (ca. 20 N).

Recent anatomical findings may help to clarify the mechanisms responsible for the greater posterior shear stiffness we measured. Tonetti et al.\textsuperscript{29} have recently reported that the anterior portions of the facet capsules are thicker and have more elastic fibres than the posterior portions. Mercer and Bogduk\textsuperscript{14} have determined that the annulus fibrosus is thicker and more fibrous anteriorly than it is laterally or postero-laterally. As these are the very structures that our results suggest transmit the posterior shear forces (Figure 5-6B) we hypothesise that it is their physical properties which may limit the posterior displacement. Preload caused greater initial shear stiffness under anterior shear than under posterior shear (Figure 5-5B). This resulted in an approximately equal shear displacement under anterior and posterior shear (Figure 5-5A). With respect to the aforementioned ligament slackening mechanism this suggests a slackening of the thickest portions of the annulus fibrosus and of the facet capsule leading to an increase in posterior displacement (Figure 5-5A).

To our knowledge this study represents the first investigation of load-sharing under shear force application in the cervical spine. The instrumentation applied allowed the clear identification of some of the load transmission paths in the cervical spine. As discussed previously, there were almost certainly load transmission paths, which we were not able to sense with the instrumentation used. This is a constraint imposed by the difficulties inherent with applying sensors while preserving as much as possible the anatomic structures and hence the segment biomechanics. It is our hope that the data presented here will be used to extend the validation of contemporary analytic models and that these models in turn can be used to analyse the load-sharing function of other anatomic structures of interest.
The bone strains exhibited considerable variation and for this reason it was not possible to apply a statistical analysis to them. We believe that this was at least partly due to variations in bone quality and vertebral anatomy. This is a limitation of our study but measuring bone strains of vertebral bone is known to be technically challenging. The variation in our data is representative of that measured by other investigators.\textsuperscript{4,23}

As with any \textit{in vitro} investigation, it was not possible to duplicate \textit{in vivo} conditions in this study. There were limitations inherent to \textit{in vitro} testing (i.e. lack of individual muscle forces, inaccuracies in the biochemical and physical environment) and to the theoretical models upon which the loading protocol were based.\textsuperscript{9,15} The results of this study must therefore be viewed as a model of the \textit{in vivo} situation, not an exact representation.

\textbf{5.5.3. CONCLUSIONS}

Although facet dislocation is generally considered a flexion\textsuperscript{30} or "distractive flexion"\textsuperscript{1} injury, our results suggest that anterior shear is also a possible mechanism of facet dislocation. Even under the non-traumatic shear magnitudes applied in this study, the facet joints are close to anterior dislocation (Figure 5-6A). Anterior shear can result from anterior acceleration of the head with respect to the torso (i.e. from impact to the back of the head or during frontal automobile collisions). Similarly, posterior shear results when the head is accelerated posteriorly with respect to the torso (i.e. from impact to the face or during whiplash). Our results suggest that tension in the anterior annulus or facet capsules may be a mechanism behind the injury to these structures, which is sometimes observed in patients after whiplash.\textsuperscript{2} Lateral shear may result from direct impact to the side of the head or from side impact automobile collisions. Our results suggest the lamina and caudal portion of the facet joint may impinge and transmit considerable forces. These structures may warrant clinical investigation after trauma associated with lateral shear.
5.6. References


5.7. Figures

Figure 5-1: Shear force application apparatus. The four IRLEDs mounted to the moulding material of each vertebra are visible. The specimen’s left side is facing the camera and its anterior aspects are at left. The specimen is hidden behind the linear bearing. Steel cables were used to apply forces to the cranial vertebra. The resulting shear force vector, $F$ is illustrated. The linear bearing was adjusted so that the shear force was applied in the mid-disc plane. The pneumatic cylinder visible above the specimen was used to apply axial preload. The preload roller allowed specimen translation.
Figure 5-2: Average principal strain magnitudes at 200 N of applied shear. The error bars indicate one standard deviation. Negative values indicate compressive strain and positive values indicate tensile strain. A) Rosette 1, left lateral mass rosette. B) Rosette 2, right lateral mass rosette. C) Rosette 3, vertebral body rosette. PL=preload.
Figure 5-3: Average change in the principal strain magnitudes at 200 N of applied shear. The error bars indicate one standard deviation. Negative values indicate compressive strain and positive values indicate tensile strain. A) Rosette 1, left lateral mass rosette. B) Rosette 2, right lateral mass rosette. C) Rosette 3, vertebral body rosette. PL=preload.
Figure 5-4: A) Average intradiscal pressure at 200 N of applied shear. B) Average change in the intradiscal pressure between 0 and 200 N of applied shear. Results for left and right shear have been combined. Error bars indicate one standard deviation. The numbers beside each error bar indicate statistically significant differences between the average pressures at P<0.05. PL=preload.
Figure 5-5: Average kinematic response for each shear force direction and preload condition. A) shear translations and B) shear stiffnesses. Results for left and right shear have been combined. Error bars represent one standard deviation. The numbers beside each error bar indicate statistically significant differences between the average value at P<0.05. PL=Preload.
Figure 5-6: Kinematics of a typical specimen (level C2-3) illustrated using reconstructed CT images. The posture of the specimen under 200 N of shear force applied in the A) anterior, B) posterior and C) left-lateral directions. A) and B) are viewed from lateral and C) from posterior. The dens has been resected from C2 to simplify moulding of the specimen. Hypothesised load-sharing mechanisms for each loading direction are illustrated.
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</tbody>
</table>

Table 5-1: Average main translations (in underlined) and coupled rotations and translations. R=rotation, T=translation, SD=standard deviation. The orientation of the coordinate system with respect to the vertebra anatomy is shown above the table.
CHAPTER 6. GENERAL DISCUSSION

The objectives of this study were to identify load-sharing mechanisms in the cervical spine under clinically-relevant loading modes. Of the 12 traditional isolated loads (flexion and extension moments, right and left lateral bending moments, right and left torsion moments, tension and compression forces, anterior and posterior shear forces, and right and left lateral shear forces), only tension was not investigated for load-sharing. Novel experimental techniques that were developed in support of this goal included:

i. A method to measure cervical disc pressure in vitro which was minimally disruptive to the specimen biomechanics (Chapter 2).

ii. The use of triaxial strain gauges to sense both the magnitude and orientation of anterior and posterior column bone strains (technique presented in Chapter 3).

iii. A graphical method to represent kinematic data from biomechanical tests (Appendix 1).

In addition, a method to apply preload during flexion testing which minimised apparatus-related artefact loads was developed (Appendix 5).

6.1. Load-sharing under compression

Our results suggest that the anterior column of the cervical spine is the major load-bearing structures under compression at both flexion and extension postures. This fact notwithstanding, the posterior column also transmitted increasing amounts of compression as the extension angle increased. Compressive transmission through the anterior column increased with increasing flexion postures and decreased with increasing extension. The posterior column usually transmitted tension (or small compression) loads under flexion postures and compression or shear (or small tension) loads under extension postures. There is evidence to support the hypothesis that the
tensile strains arose from tension in the facet capsules. We assume compressive strains were transmitted through the anterior annulus anteriorly and through contact at the facet joints posteriorly. The compressive force transmission through the posterior column resulted in measurable strains at the lateral mass rosettes. However, tensile strains were present even under some extension postures (Figure 3-6) and there was a consistently large magnitude of compressive strain and invariant disc pressure at the anterior column. It is based on these facts that we conclude that the bulk of the compressive force is flowing through the anterior column even under extension postures.

Our results were in good agreement with those of Pintar et al. under flexion loading where the applied loads were similar. Under extension loading, the results diverged. In the present study, we found compressive force transfer at the anterior column while Pintar et al. measured tension loading. This was likely due to the extension moment associated with the posterior eccentric compressive load applied by Pintar et al. Our results were also in accordance with the theoretical results of Goel and Clausen who found that 88% of the applied compressive force in the neutral posture is transmitted through the anterior column.

The compressive vertebral body strains were consistently oriented within 30° of the spine’s cranial-caudal axis. The lateral mass principal strains exhibited more variation in orientation. Strains at both locations varied considerably in magnitude. Other investigators have measured similar variation in vertebral bone strains when using strain gauges for measurement. We believe the variation in bone strain was caused by inter- and intra-specimen variability in vertebral anatomy and bone quality. Difficulties inherent in reproducing the rosette location on each of the lateral masses for specific specimens and across specimens also contributed to the high strain variability. The strain variation rendered statistical analysis with the present number of specimens impractical.
The observed variation in principal strain orientations suggests that the application of triaxial strain rosettes was necessary to accurately characterise the bone strains. Uniaxial gauges would have sensed lower strain magnitudes than those actually present for strains not aligned with gauge. Our inferior reaction load data confirmed that no large extraneous loads were associated with our load application apparatus.

Our results support the hypothesis that injurious magnitudes of compressive load may be transmitted through the posterior column because of neck muscle activity during the neck extension phase of whiplash. In this scenario, the neck musculature is assumed to become active during neck extension after the rear-end collision. This is believed to cause a compressive force of injurious magnitude to be applied to the spine which, owing to the extended spine posture, is transmitted through the facet joints leading to injury of these structures. Recent experimental and clinical evidence supports this theory. *In vivo* vertebral motion data from a volunteer study suggest the existence of potentially injurious facet “impingement” during whiplash. The hypothesis that neck muscles become active during neck extension has been confirmed using electromyographic sensors. There is also mounting clinical evidence that cervical facet joints are the source of pain for many patients suffering from chronic neck pain subsequent to whiplash. Further work is necessary to establish what magnitudes of compression correspond to facet joint injury under extension postures.

Cervical disc pressure increased linearly with applied compressive force and was approximately invariant with respect to flexion and extension postures. This reflected the disc's central role in compressive load-sharing at all flexion-extension postures tested. Extrapolating slightly from our results, pressures of approximately 4 MPa are expected in the cervical disc under loads of 1000 N. This is approximately 400% greater than the corresponding pressure known to occur in lumbar discs under the same load.
To our knowledge, this is the first time that cervical intervertebral disc pressure has been measured as a function of compression loading. Despite two preconditioning cycles and a 30-second creep period at the maximum applied load, the disc pressure was not repeatable. It decreased as the number of tests increased. This suggests that load may have to be applied over a longer period before the disc creeps to a state of repeatable equilibrium. Such an approach has been advocated for the lumbar spine by Adams but it is not routinely applied by researchers active in lumbar spine biomechanics.

The sequence-associated change in disc pressure was small (approximately 15%) compared to the absolute disc pressures measured under 800 N of compression. This implies that large posture dependant changes in disc pressure (i.e. >50%) would have been detected despite the non-repeatability of disc pressure. This was our rationale for concluding that disc pressure was approximately invariant with changes in posture under this protocol.

6.2. Load-sharing under applied bending moments

Distinct load-sharing mechanisms were identified for each of the flexion-extension, lateral bending and axial torsion loading modes. Flexion moments were supported by compression at the anterior column and tension at the posterior column. This is consistent with that expected based on elementary beam theory. Under extension moments, the opposite was true, compression was transmitted posteriorly and tension anteriorly. There is evidence to suggest that the tensile forces arose from tension in the facet capsules posteriorly. We assume anterior tension arises from tension in the disc annulus and anterior longitudinal ligament. Compressive forces are likely transmitted through the disc annulus anteriorly and via facet contact posteriorly. Under these loading conditions, strains were usually aligned with the cranial-caudal axis of the spine.
Under lateral bending loads, the force was transmitted through the anterior and posterior columns in two distinct manners. The facet joint on the side toward which the bending moment was applied transmitted increasing amounts of force and the opposite facet joint transmitted decreasing amounts of force. The bone strains were typical of shear strain, not of tension or compression as they were under flexion and extension moment application. In contrast to the flexion-extension results, tensile strains were not detected at the less-loaded lateral mass under lateral bending. The disc was deformed under this loading mode which was manifested as increases in disc pressure and vertebral body strain. We hypothesise that the anterior column supported some of the applied moment in resisting this deformation.

Axial torsion moments were also transmitted through both anterior and posterior columns. In contrast to the lateral bending case, the posterior column load-sharing was approximately equal at both lateral masses independent of the direction of the applied torsion. This was surprising since intuitively we expected the facet joint opposite to the direction of the applied torsion to be forced into contact (i.e. right facet joint contact for left lateral bending) and thus transmit more force. We hypothesise that the large coupled lateral bending associated with torsional moment was responsible for this. As the torsional motion brought the facets contralateral to the moment application closer together, the coupled lateral bending motion moved them further apart. In a similar manner to lateral bending, the disc was deformed in response to the applied moment and thus we believe it supported some of the applied load.

Similar intervertebral pressure increases were recorded under flexion and extension loading. This was surprising since intuitively we expect tension at the vertebral body gauge and in the anterior annulus to be indicative of an unloaded disc (i.e. less pressure). We hypothesise that the cranial vertebra “pivoted” around the taught annulus anteriorly causing a net compression to be applied to the disc. Our kinematic results
support this hypothesis. The HAM between 0 and 0.5 Nm was located approximately at
the anterior annulus. It moved posteriorly between 0.5 and 1.0 Nm. This caused the
overall 0-1 Nm HAM to be located at the anterior spinal canal. The disc height data also
indicated net disc compression under both flexion and extension.

Under all loading modes, axial preload was transmitted primarily as compressive
force in the anterior column. Preload did not change the lateral mass load-sharing. This
finding is consistent with the compression load-sharing component of this study.

To our knowledge, no previous experimental data exists about load-sharing in
the cervical-spine under isolated bending moment application. Our results compare well
with Goel and Clausen’s theoretical model except under axial torsion where they
predicted an unloaded ipsilateral facet and a highly loaded contralateral facet. In
contrast, we found equal load-sharing at both lateral masses. Our disc pressure data
was in general agreement with the in vitro data from Pospiech et al.\(^7\) who also reported
increases in disc pressure with applied extension moments. Our preloaded disc pressure
data was also similar to the in vivo data reported by Hattori et al.\(^8\) Our values were
typically bracketed by the in vivo data for flexion and extension postures. This indicates
that the 200 N preload was adequate to represent non-traumatic active neck
musculature in vivo.

The variation in bone strain data measured in this study was caused by the same
factors discussed earlier (i.e. anatomical variation and difficulties inherent in
standardising rosette placement). The variation also rendered statistical analysis of bone
strain data impractical for this experiment. Except under flexion-extension loading the
strain orientations were irregular and no pattern could be discerned. This was likely a
result of the strong coupling under other loading modes. The bone strains arose from
both bone contact and ligament and annulus tension. The lateral bending-axial rotation
coupling resulted in complicated patterns of relative motion between relevant bone and
ligament attachment points. HAM information proved to have excellent utility for the analysis of load-sharing. For example, the subtle difference between coupling characteristics when torsion moments were applied versus when lateral bending moments were applied was identified using the HAM data.

The non-repeatability observed in the compressive study raises the question of whether similar phenomena affected the disc pressure under flexion or shear loading. No formalised investigation of disc pressure repeatability was undertaken in this study. However, for one specimen, data was collected for all three loading cycles (two preconditioning and one measurement). This was done to evaluate if two preconditioning cycles were sufficient. The results suggested that the disc pressure was repeatable over the three cycles. We hypothesise that the disc does not creep as much during these tests due to the lower loads applied.

Our data suggest compressive forces are primarily transmitted through the anterior column and the facet joints act as “guides” to the segment motions. Our results identify tissues that may be highly loaded for each of the loading modes. These tissues may be at risk of failure under traumatic magnitudes of moment. Therefore, our results suggest particular anatomic structures that could be regarded as preferentially at risk of injury at clinical examination. For example, facet joints may be at risk of compressive injury under traumatic extension moments while posterior facet capsule rupture may be suspected with flexion-associated trauma.

6.3. Load-sharing under applied shear loads

Under applied shear loads, cervical FSUs exhibited distinct load-sharing mechanisms that were dependent on the shear direction. Under anterior shear, the cranial facet joints ramped up the caudal facet joints, which caused a coupled flexion rotation. As anterior shear forces increased, the cranial vertebra pivoted about the compressed disc annulus anteriorly and continued to ramp up the facet joints posteriorly.
The facet joint contact resulted in net compressive or shear strains at the lateral mass rosettes. Together the anterior pivoting and posterior ramping caused a relative decrease in disc pressure under anterior shear.

Under posterior shear, a posterior translation was measured with little associated kinematic coupling. The disc was therefore subjected to a primarily shear deformation. This was manifested as an increase in disc pressure and tension at the vertebral body rosette. The lateral mass rosettes exhibited compressive strain, which was surprising, since the animation data suggested that the facet joints were distracted. We hypothesise that the posterior shear translation resulted in tension in the anterior portion of the facet joint capsule. We believe the tension force resulted in a bending moment being applied to the lateral mass, which would explain the compressive strain at the lateral mass rosettes. The tension measured at the vertebral body was likely transmitted through the anterior annulus and anterior longitudinal ligament.

Lateral shear was associated with coupled lateral bending and axial torsion rotations. Shear strain was measured at all rosette locations. The net disc deformation caused by the lateral shear deformation and the coupled rotations caused an increase in disc pressure. We expected one of the lateral masses to be more highly loaded depending on the direction of the lateral shear force but this was not the case. The animation and kinematic data suggested that the superior lateral mass impinged against the inferior lamina on the side opposite the direction of shear force application. We believe that the lack of a direction-dependent increase in strain at the lateral masses may indicate that large forces are being transmitted through the bony impingement. This may have the effect of decreasing the resulting lateral mass strains. The coupled rotations may also play a role in harmonising the lateral mass strains under lateral shear. As described above, the lateral bending and axial torsion result in competing effects on the facet joint contact. Thus we hypothesise that the shear strains at all rosette locations
were likely transmitted through the taught disc annulus anteriorly and facet capsules posteriorly. Owing to the lateral translation, the annulus and facet capsule forces were directed somewhat laterally which resulted in shear strain at the rosette locations. In addition, our results suggest that force is also transmitted through the facet joint-lamina impingement.

As in the other experiments, the bulk of the compressive preload force was transmitted through the anterior column. This was manifested as increased disc pressure and vertebral body compressive strain. Preload caused the posterior column load-sharing to remain unchanged or to moderately decrease. This behaviour again suggests that the facet joints act primarily to guide segment motion and play a relatively small role in compressive load transmission.

The anterior shear stiffness was lower than the posterior shear stiffness. This resulted in a greater anterior shear translation compared to posterior shear. This is a surprising result since many authors have hypothesised that the facet joints provide a physical stop to anterior displacement.\textsuperscript{10,21} In addition, the lumbar spine has been found to be stiffer under anterior shear than under posterior shear.\textsuperscript{21} Facet geometry is one possible explanation for this regional difference. In the lumbar spine, the facets are more vertically oriented.\textsuperscript{18} This suggests a physical block to anterior translation as opposed to the ramp-like geometry present in the cervical spine. Recent anatomical findings suggest additional morphological bases for the increased posterior stiffness. Both the facet capsules\textsuperscript{22} and the disc annulus\textsuperscript{14} have been reported to have thicker and more fibrous portions anteriorly than posteriorly or laterally. These are exactly the structures that our results indicate transmit tensile force under posterior shear. Since larger dimensions and more fibres generally indicate higher tissue stiffness these structures likely contribute to the difference in anterior and posterior shear stiffness.
Preload caused a decrease in stiffness for posterior, lateral and anterior (180-200 N) directed shear. Under posterior and lateral shear the decreased stiffness corresponded to significantly greater shear translations. This behaviour contradicts preload-associated increases in shear stiffness in the lumbar spine\textsuperscript{10,21} and increases in bending stiffness in the lumbar (Appendix 5) and cervical (Chapter 4) spine. Yang and Begeman\textsuperscript{24} have proposed a mechanism for the preload-associated decrease in stiffness. They hypothesise that straightening of the thoracic spine’s kyphosis during whiplash results in cervical spine compression. This is believed to cause the intervertebral ligaments to slaken. The FSU with slackened ligaments is believed to exhibit decreased shear stiffness. Our results confirm this hypothesis for lateral and posterior shear.

The decrease in disc pressure under anterior shear has not been reported previously at any spinal region. In the lumbar spine, the disc pressure is known to increase under anterior shear loading.\textsuperscript{4,6} We hypothesise that the previously described difference in facet geometry may be responsible for this discrepancy as well. In the lumbar spine the vertically oriented facets do not promote the posterior ramping under anterior shear that we observed with our specimens.

To our knowledge, this study represents the first experimental evaluation of load-sharing in the cervical spine under shear loading. This experiment suffered from the same limitations with respect to anatomical variation and difficulties inherent in duplicating the sensor placement discussed earlier. The variation in bone strain data rendered statistical analysis with the present number of specimens impractical. Some shear load-sharing mechanisms may have been present which we could not detect with our instrumentation technique. For example, we did not detect the force flowing through the facet-lamina impingement with any of our strain rosettes. We could also not determine if the ramping observed under anterior shear resulted in tension at the
posterior annulus. Data from one specimen's preconditioning cycles suggest that the disc pressure was repeatable under this loading mode.

The use of highly accurate motion analysis system and standardised anatomy-based local co-ordinate systems allowed small kinematic difference in stiffness and displacement to be identified. Pilot studies that we undertook indicated that this behaviour would not have been detectable using actuator displacement data from the materials testing machine.

The animation system allowed the identification of important load-sharing mechanisms (i.e. facet-lamina impingement under lateral shear). These mechanisms would have been undetectable with conventional kinematic data representations.

Our results suggest distinct load-sharing mechanisms for each of anterior, posterior and lateral shear. The specific anatomic structures that were shown to transmit relatively high forces could be preferentially examined for patients known to have undergone traumatic instances of these loads. For example under whiplash loading the cervical spine is likely subjected to posterior shear due to inertial loading by the head as the torso is accelerated. Our results suggest that the anterior annulus and anterior facet capsules may be at risk of tension injury under these loads. This is consistent with the injuries observed clinically for some patients after whiplash. Our results confirm the preload-associated stiffness decrease hypothesised by Yang and Begeman under quasi-static conditions. Further work is necessary to examine if this behaviour is present at the dynamic load rates present during whiplash.

6.4. Test duration

The complete series of tests (compressive, flexibility and shear) were performed over two testing days of approximately 10 hours each. Extreme care was taken to prevent the specimens from dehydrating. FSU biomechanics have been shown not to significantly degrade under these conditions.
6.5. References


CHAPTER 7. SUMMARY AND CONCLUSIONS

Load-sharing mechanisms in the human spine have been identified under all traditional isolated loading modes except tension. Under compression loading, the influence of superimposed flexion and extension postures was established. Under all other loading modes the effect of a constant axial preload was evaluated. The preload represented an active neck musculature and the weight of the head.

In support of the load-sharing experiments novel techniques were developed to measure cervical intradiscal pressure, and animate kinematic results. In addition, a technique was developed and applied which minimised the load artefacts associated with preload application during flexibility tests.

7.1. General conclusions

The results of these investigations allow the following conclusions that were evident under all loading modes:

- The methods used allowed a qualitative evaluation of the load passing through the anterior and posterior columns of the spine. Quantitative evaluation of the loads transmitted through individual anatomic components was beyond the capabilities of the present methods.
- Under all loading modes the bulk of applied compression loads are transmitted through the anterior column. The lateral masses appear to function primarily as "guides" for the motion that is produced when non-compressive loads are applied.
- Vertebral bone strain magnitudes and orientation are indicative of the loading mode. They also exhibit large inter- and intra- specimen variation. Therefore, the application of triaxial rosettes is necessary to avoid underestimating the true strain state.
7.2. Compression load-sharing

The compressive load-sharing study supported the following conclusions:

- Cervical disc pressure increases linearly with applied compression.

- For the same applied compression cervical disc pressure is approximately four times higher than lumbar disc pressure. An approximate “rule of thumb” is that cervical disc pressure increases 4 MPa and lumbar disc pressure 1 MPa per 1000 N of compression.

- Under compression loading, cervical disc pressure varies little with changes in flexion-extension posture.

- Increasing flexion postures cause increasing compressive force transmission through the anterior column and decreasing compressive transmission or tension transmission through the posterior column.

- Increasing extension angles result in increasing compressive force transmission through the posterior column and decreasing compressive force transmission through the anterior column.

- Large compressive forces may be transmitted through the posterior column under compression loading at large extension angles.

7.3. Moment load-sharing

The applied moment load-sharing study supported the following conclusions:

- Flexion moments result in compressive force in the anterior column and tensile force in the posterior column. Under extension, the opposite is true.
• Lateral bending moments are supported by internal loads in the disc, increased load at the facet toward which the bending was applied and decreased load at the opposite facet.

• Torsion moments result in internal loads in the disc and equal loads at both facet joints.

• Flexion and extension moments result in equal increases in disc pressure. For extension loading, this suggests a mechanism whereby the anterior annulus and anterior longitudinal ligament are loaded in tension. The cranial vertebra appears to pivot about the taught annulus and ligament resulting in disc compression. The disc pressure caused by lateral bending and torsion moments was one half and one third respectively of that for flexion and extension loading.

7.4. Shear load-sharing

The shear loading study suggests the following conclusions:

• Under anterior shear, the cranial vertebra pivots about the compressed disc annulus anteriorly and the facet joints slide up the caudal facet joint surfaces posteriorly. Together these factors result in a coupled flexion rotation and a decrease in disc pressure.

• Posterior shear forces are supported by internal forces in the disc and tensile forces at the vertebral body. Bending moments may be applied to the lateral masses through tensile forces in the anterior facet capsule.

• Lateral shear forces are supported by internal forces in the disc and shear force transmission at the vertebral body and lateral masses. In some instances, the cranial facet joint opposite to the direction of the applied shear force impinges against the underlying lamina. Force may also be transmitted through this impingement.
Posterior shear stiffness is higher than anterior shear stiffness.

Compressive preload results in decreased stiffness under posterior, lateral, and anterior (between 180 and 200 N of applied shear) shear.

7.5. Recommendations

Based on the results of these studies we make the following recommendations for future work:

- More work is necessary to determine if extreme flexion or extension angles cause significant changes in disc pressure under compression loading. Conditions under which repeatable disc pressures can be measured for compression loading have to be established to investigate this question.

- The investigation of load-sharing mechanisms at more extreme cervical spine postures typical of those observed in vivo is of interest. However, if non-destructive loads are intended steps must be taken to ensure the anatomic specimens (typically from elderly donors) are not injured.

- Load-sharing mechanisms and their variation as a function of destructive loads would be of interest and have immediate applicability to the identification of mechanisms of injury seen clinically.

- Load-sharing under non-destructive and destructive dynamic loads typical of trauma (e.g. whiplash) is of interest and would also be applicable for the identification of injury mechanisms.

- Our results could be used to validate finite element or other theoretical models of the cervical spine against paired load-sharing and kinematic data.

- Because facet capsules are suspected as a source of neck pain, direct measurement of facet capsule strain is recommended for future studies.
- Our results suggest that additional load transmission paths exist. Rosettes mounted on the posterior vertebral body and on the lamina may allow identification of additional important load transmission paths.

- Upper cervical spine (C2-occiput) load-sharing could be addressed using a similar instrumentation technique to the one applied here (not including the disc pressure sensor since there is no disc at this level.)
Appendices
APPENDIX 1. ANIMATION OF IN VITRO 
BIOMECHANICAL TESTS

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A1.1. Introduction

In the interdisciplinary world of orthopaedic biomechanical research, it is often difficult to communicate complex kinematic and load concepts so that it is understandable by all members of the interacting disciplines, such as surgeons, engineers, biologists etc. The objectives of this study were to develop a method to graphically replay in vitro biomechanical tests. This will allow intuitive interpretation of kinematic and force data independent of the viewer’s area of expertise.

A1.2. Materials and Methods

A level C2-3 human cervical spine segment consisting of two vertebrae and their intervertebral disc and ligaments was moulded in polymethylmethacrylate (PMMA) blocks as described in Chapter 3 (Figure A 1-1). Three aluminium spheres (diameter=3.0 mm) were glued to each vertebra and used as fiducial markers (Figure A 1-1 and Figure A 1-2). The specimen with attached markers was scanned using computer tomography (CT) with slice thickness of 1.0 mm and slice spacing of 1.5 mm.

IRLED markers were attached to each vertebra (Figure A 1-1). An optoelectronic measurement system (Optotrak 3020, Northern digital, Waterloo Canada) was used to
record the specimen kinematics during the biomechanical tests described in Chapters 3-5. Aluminium sphere locations were measured using an Optotrak-based space pointer (Figure A 1-1). This data was used to "match" the CT-based model motion to the experimental specimen motion.

Three-dimensional models of the C2 and C3 vertebra were produced by segmenting and reconstructing the CT data (Figure A 1-2). The biomechanical tests were replayed using the three-dimensional CT models and a Silicon Graphics workstation. The animation program allowed the specimens to be viewed from any angle or distance. HAM data, calculated as described in Chapter 4, could also be rendered in the animation. Custom software developed by the Computer-Aided-Surgery research group at the Müller Institute was used for the CT reconstruction, aluminium marker-CT model matching and graphical playback.

**A1.3. Results**

Matching errors (sum of the distances between the CT and the space pointer markers) were less than 1.0 mm RMS for both vertebrae. A typical animation frame for a left lateral bending moment with preload is illustrated in Figure A 1-3 (from three different perspectives).

HAM data could not be animated continuously. A minimum angle of approximately 0.25 degrees was necessary before a meaningful HAM could be calculated.

**A1.4. Discussion**

The animated tests allowed synthesis of the disparate specimen kinematic and HAM data and aided in the interpretation of the load-sharing data (Chapters 4 and 5). The viewer could move, rotate and magnify the CT models as desired. It was thus possible to examine specific regions of the anatomy in detail during test playback. For
example, it was possible to examine the location of facet contact and the variation of
disc height as the test progressed. In some instances (i.e. under lateral shear, Chapter
5) other bony contact was identified.

It was not possible to animate the HAM continuously. It is well established that
specimen translations must be minimal and a finite rotation must be present to calculate
a meaningful HAM.\(^1\) However, even with this restriction we were typically able to
calculate 6 to 12 HAM per flexibility test. As discussed previously (Chapter 4), the ability
to examine intermediate HAM —especially simultaneously with kinematic playback—
allowed much more detailed analysis of the motion than average HAM data did.

The vertebrae were assumed rigid, thus deformations were not reflected in the
animations. This could lead to unrealistic animations at higher load levels. This
technique improves the ability of experts from disparate backgrounds to interpret and
discuss this type of biomechanical data.

A1.5. References

   estimation from noisy landmark measurements in the study of human joint
A1.6. Figures

Figure A 1-1: Identification of the aluminium sphere locations. A close-up of the sphere digitising probe is shown in the insert.

Figure A 1-2: Location of the aluminium spheres on the CT models. Left - C2 from inferior. Right - C3 from superior.

Figure A 1-3: Animation of the specimen and HAM at under a left lateral bending with preload. The dens has been transected.
APPENDIX 2. SUPPORTING DATA FOR LOAD-SHARING UNDER COMPRESSION (CHAPTER 3)

The following data further supports the conclusions made in Chapter 3. In the first section, details of the bone strain are given. Specifics of the shear-dominated lateral masses, strain orientations and the range of the principal strain data are presented.

A2.1. Bone Strain

Figure A 2-1: Average normalised strain plotted versus posture for the lateral mass rosettes that were dominated by shear strain.

Figure A 2-2: Orientation of the principal strains at the vertebral body.
Figure A 2-3: Orientation of the principal strains for compressive-dominated lateral mass rosettes.

Figure A 2-4: Orientation of the principal strains for tensile-dominated lateral mass rosettes.

Figure A 2-5: Individual specimen compressive principal strain magnitudes at the vertebral body. Each symbol represents one of the 6 specimens.
Figure A 2-6: Individual specimen tensile principal strain magnitudes. Each symbol represents one of the 12 lateral masses.

Figure A 2-7: Individual specimen compressive principal strain magnitudes. Each symbol represents one of the 12 lateral masses.
A2.2. Disc Pressure

In this section a representative example of the strong effect that test sequence had on the disc pressure results is presented.

![Disc Pressure Graph](image)

Figure A 2-8: Disc pressure repeatability for a representative C3-4 specimen. Each pressure curve is indicated by the posture. The bracketed number indicates the tested sequences. (i.e. 1 is the first posture tested and 10 represents the 10 posture tested)

A2.3. Kinematics

In this section, the disc-height sequence-dependence is presented.

![Disc Height Chart](image)

Figure A 2-9: Dependence of disc height on sequence at the neutral posture. The difference between disc heights at 800 N of applied load as a function of the number of tests performed (sequence difference) is presented. Each data point represents the results from one specimen. Some specimens were tested at the neutral posture three times.
A2.4. Reaction Loads

In this section, the average reaction forces are presented. See Figure 3-3 for the co-ordinate system orientation.

Figure A 2-10: Average reaction force data as a function of flexion-extension posture. \( F_y \) is the vertical reaction force.
APPENDIX 3. SUPPORTING DATA FOR LOAD-SHARING UNDER APPLIED MOMENTS (CHAPTER 4)

The following data further supports the conclusions made in Chapter 4. In the first section, average strain values and their variation with moment are presented for all loading cases and rosette locations that were not presented in the chapter.

A3.1. Bone strain

![Figure A 3-1: Average strain magnitude and orientation for the left lateral mass under flexion-extension.](image1)

![Figure A 3-2: Average strain magnitude and orientation for the right lateral mass under flexion-extension.](image2)
Figure A 3-3: Average strain magnitude and orientation for the left lateral mass under flexion-extension with preload.

Figure A 3-4: Average strain magnitude and orientation for the right lateral mass under flexion-extension with preload.

Figure A 3-5: Average strain magnitude and orientation for the left lateral mass under lateral bending.
Figure A 3-6: Average strain magnitude and orientation for the right lateral mass under lateral bending.

Figure A 3-7: Average strain magnitude and orientation for the vertebral body under lateral bending.

Figure A 3-8: Average strain magnitude and orientation for the left lateral mass under lateral bending with preload.
Figure A 3-9: Average strain magnitude and orientation for the right lateral mass under lateral bending with preload.

Figure A 3-10: Average strain magnitude and orientation for the vertebral body under lateral bending with preload.

Figure A 3-11: Average strain magnitude and orientation for the left lateral mass under torsion.
Figure A 3-12: Average strain magnitude and orientation for the right lateral mass under torsion.

Figure A 3-13: Average strain magnitude and orientation for the vertebral body under torsion.

Figure A 3-14: Average strain magnitude and orientation for the left lateral mass under torsion with preload.
A3.2. Disc Pressure

In this section the average disc pressure behaviour as a function of moment application is given. Data for all loading directions and preload conditions is presented.
Figure A 3-17: Average variation in disc pressure under flexion-extension moments.

Figure A 3-18: Average variation in disc pressure under Lateral bending moments.

Figure A 3-19: Average variation in disc pressure under torsion moments.
Figure A 3-20: Average variation in disc pressure under flexion-extension moments with preload.

Figure A 3-21: Average variation in disc pressure under lateral bending moments with preload.

Figure A 3-22: Average variation in disc pressure under torsion moments with preload.
A3.3. Kinematics

In this section, the average rotational motion is presented, in terms of Euler angles, for each loading case.

Figure A 3-23: Average rotation under flexion-extension moments.

Figure A 3-24: Average rotation under flexion-extension moments.
Figure A 3-25: Average rotation under torsion moments.

Figure A 3-26: Average rotation under flexion-extension moments with preload.

Figure A 3-27: Average rotation under lateral bending moments with preload.
Figure A 3-28: Average rotation under torsion moments with preload.
APPENDIX 4. SUPPORTING DATA FOR LOAD-SHARING UNDER SHEAR (CHAPTER 5)

The following data further supports the conclusions made in Chapter 5. In the first section, details of the bone strain are given. Because the behaviour in left and right shear was typically symmetric, only left shear, posterior and anterior data is presented below. Data is presented for each rosette location.

A4.1. Bone strain

Figure A 4-1: Average principal strain magnitude and orientation at the left lateral mass under anterior shear.

Figure A 4-2: Average principal strain magnitude and orientation at the right lateral mass under anterior shear.
Figure A 4-3: Average principal strain magnitude and orientation at the vertebral body under anterior shear.

Figure A 4-4: Average principal strain magnitude and orientation at the left lateral mass under anterior shear with preload.

Figure A 4-5: Average principal strain magnitude and orientation at the right lateral mass under anterior shear with preload.
Figure A 4-6: Average principal strain magnitude and orientation at the vertebral body under anterior shear with preload.

Figure A 4-7: Average principal strain magnitude and orientation at the left lateral mass under posterior shear.

Figure A 4-8: Average principal strain magnitude and orientation at the right lateral mass under posterior shear.
Figure A 4-9: Average principal strain magnitude and orientation at the vertebral body under posterior shear.

Figure A 4-10: Average principal strain magnitude and orientation at the left lateral mass under posterior shear with preload.

Figure A 4-11: Average principal strain magnitude and orientation at the right lateral mass under posterior shear with preload.
Figure A 4-12: Average principal strain magnitude and orientation at the vertebral body under posterior shear with preload.

Figure A 4-13: Average principal strain magnitude and orientation at the left lateral mass under left shear.

Figure A 4-14: Average principal strain magnitude and orientation at the right lateral mass under left shear.
Figure A 4-15: Average principal strain magnitude and orientation at the vertebral body under left shear.

Figure A 4-16: Average principal strain magnitude and orientation at the left lateral mass under left shear.

Figure A 4-17: Average principal strain magnitude and orientation at the right lateral mass under left shear.
Figure A 4-18: Average principal strain magnitude and orientation at the vertebral body under left shear.

A4.2. Disc Pressure

In this section, the variation in disc pressure as a function of shear force is presented.

Figure A 4-19: Average disc pressure as a function of shear force under anterior shear.
Figure A 4-20: Average disc pressure as a function of shear force under posterior shear.

Figure A 4-21: Average disc pressure as a function of shear force under left shear.

Figure A 4-22: Average disc pressure as a function of shear force under anterior shear with preload.
Figure A 4-23: Average disc pressure as a function of shear force under posterior shear with preload.

Figure A 4-24: Average disc pressure as a function of shear force under left shear with preload.

**A4.3. Kinematics**

In this section, the average translation data as a function of shear force is presented.
Figure A 4-25: Average translation as a function of shear force under anterior shear.

Figure A 4-26: Average translation as a function of shear force under posterior shear.

Figure A 4-27: Average translation as a function of shear force under left shear.
Figure A 4-28: Average translation as a function of shear force under anterior shear with preload.

Figure A 4-29: Average translation as a function of shear force under posterior shear with preload.

Figure A 4-30: Average translation as a function of shear force under left shear with preload.
APPENDIX 5. IN VITRO AXIAL PRELOAD APPLICATION
DURING SPINE FLEXIBILITY TESTING: TOWARDS
REDUCED APPARATUS-RELATED ARTEFACTS

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A5.1. Abstract

Presently there is little consensus about how, or even if, axial preload should be
incorporated in spine flexibility tests in order to simulate the compressive loads naturally
present in vivo. Some preload application methods are suspected of producing
unwanted "artefact" forces as the specimen rotates and, in doing so, influencing the
resulting kinematics. The objective of this study was to quantitatively compare four
distinct types of preload which have roots in contemporary experimental practice. The
specific quantities compared were the reaction moments and forces resulting at the
intervertebral disc and specimen kinematics. The preload types incorporated increasing
amounts of caudal constraint on the preload application vector ranging from an
unconstrained dead-load arrangement to an apparatus that allowed the vector to follow
rotations of the specimen. Six human cadaveric spine specimens were tested. Pure
moments were applied to the specimens with each of the four different types of compressive preload. Kinematic response was measured using an opto-electronic motion analysis system. A six-axis load cell was used to measure reaction forces and moments. Artefact reaction moments and shear forces were significantly affected by preload application method and magnitude. Unconstrained preload methods produced high artefact moments and low artefact shear forces while more constrained methods did the opposite. The kinematic response parameters were highly correlated to the extra moments and forces induced. A mechanical trade off is suggested by our results, whereby unwanted moment can only be prevented at the cost of shear force production. Caution should be exercised when designing or interpreting spine flexibility studies including axial preload. Unwanted moments or forces induced as a result of preload application method may render the comparison of two seemingly similar studies inappropriate.

A5.2. Introduction

In vitro characterisation of flexibility is widely used to investigate the natural behaviour of the spine and the effect of surgical techniques and devices. Presently there is little consensus about how, or even if, axial preload should be used in order to simulate the compressive loads naturally present in vivo.¹⁷

Numerous investigators¹⁻⁵,⁷⁻¹²,¹⁴,¹⁵,²¹ have concluded that preload significantly affects the flexibility of the spine, although there is widespread disagreement as to the quantity and quality of this effect. Compressive preload application has also been suspected of inducing additional, unwanted bending moments.⁶,⁷,¹¹,²² Although the extent of this effect has never been conclusively established, a variety of preload application methods have been devised.

Panjabi et al.¹² fixed a hanging weight above the cranial vertebra and found that specimen flexibility increased in flexion and lateral bending. Miller et al.¹¹ and Janevic et
al. argued that this set-up would encourage large additional bending moments and hence they lowered their upper fixation point to the geometric centre of the cranial vertebra. Janevic et al. further constrained the displacement of the hanging load with rotation, by fixing the cable to a stationary point below the specimen. In contrast to Panjabi et al. both groups found that flexibility in flexion and lateral bending had decreased. In extension where Panjabi et al. had found no effect, Janevic et al. again found a decrease, while Miller et al. found an increase. In our laboratory an attempt has been made to further minimise the effect of rotation on the movement of the preload application vector. The length of the cable was reduced by bringing the fixation points closer to the disc centre. Using a preload of 600 N, preload was found to have no effect on flexion, extension or lateral bending kinematics in the intact lumbar spine. Patwardhan et al. recently reported that a “follower load”, consisting of a compressive force vector which joins the geometric centres of adjacent vertebral bodies significantly stabilises the lumbar spine allowing it to carry more compressive load before buckling. There is general agreement for axial torsion, where additional moments are not believed to be induced, that flexibility decreases with preload application.

Although numerous authors have referred to a preload-associated increase in applied moment, to date only comparisons of change in kinematic response have been reported. A study which examines the moments and forces, induced as a result of preload application, and their relationship to the resulting kinematics is needed.

The objective of this study was to quantitatively compare four distinct types of preload which are commonly used in contemporary experimental practice. The specific quantities compared were the reaction moments and forces resulting at the intervertebral disc and specimen kinematics. A further objective was to develop a mathematical model of preload application methods suitable for relative comparison of preload configurations which were not evaluated in the experimental portion of this study.
A5.3. Materials and Methods

A5.3.1. SPECIMEN PREPARATION

Six lumbar functional spine units (FSU) were harvested from different donors, screened radiographically for bony abnormalities and frozen until use. The segment used (1-L1/L2, 3-L2/L3, 1-L3/L4, 1-L4/L5) was prepared in accordance with accepted procedures\textsuperscript{20} and moulded into rectangular blocks of polymethylmethacrylate (PMMA) such that the mid-disc plane was oriented in the horizontal plane (Figure A 5-1). The specimens were mounted in a custom testing machine which used bearing-mounted pneumatic cylinders and steel cables to apply pure, non-constraining moments (accurate to within ± 0.1 Nm, Figure A 5-1).

A5.3.2. LOADLING APPARATUS

The preload was positioned on the upper PMMA block so that the maximum (400 N) load did not produce observable rotations, away from the neutral posture, in any plane. This is often referred to as the balance point of the specimen.\textsuperscript{16,19} The preload was applied via cables, whose attachment points were adjustable so that the preload could be fine tuned to the balance point in the sagittal and frontal planes.

The four types of preload were:

Type I: was fixed to the rotating body, remained constant in magnitude, and its direction was preserved (vertically due to gravity). Experimentally, this consisted of a free hanging weight, hung directly below the specimen. It was fixed only to a point above the cranial vertebra and was unconstrained in all six degrees of freedom (Figure A 5-2A and Figure A 5-3A).

Type II: was a modification of Type I by adding a lower constraint on the preload cables. Cylindrical guides, with openings slightly larger than the diameter of the wire, were
attached below the caudal vertebra at the base of the load cell (Figure A 5-2B and Figure A 5-3B).

Type III: this type was a modification of Type II, with the upper and lower fixation points now level with the edges of the intervertebral disc. To achieve this, extended cylindrical guides were attached to the guided rod above the cranial vertebra and to the lower guides described in Type II (Figure A 5-2C and Figure A 5-3B).

Type IV: was an experimental interpretation of a follower load, whereby the constant preload was kept perpendicular to the end plates of the cranial vertebra during the rotation using a specially designed device. The lower guide position could be adjusted antero-posteriorly without disconnecting the preload. It was adjusted following each moment application step to realign the preload cables to a guide fixed to the upper vertebra (Figure A 5-2D and Figure A 5-3C). To simulate continuous motion, the number of loading steps for this type was doubled. It was not possible to use this preload set-up for axial torsion tests.

A5.3.3. TEST PROTOCOL

Preload was applied by adding weights of 20 and 40 kg. Due to differences in the apparatus for each type the actual preload force varied slightly. The load magnitudes will be consistently referred to as either 200 or 400 N.

The origin of the three-dimensional co-ordinate system was placed at the centre of the intervertebral disc (Figure A 5-1). The co-ordinate system was fixed to the caudal vertebra and thus did not rotate with the cranial vertebra. One full flexibility test consisted of applying pure moments of flexion-extension, bilateral axial rotation, and bilateral lateral bending individually, to a maximum of 5.0 Nm, in four equal steps (eight equal steps for Type IV), for two cycles. The first cycle was used for preconditioning and no data from it was analysed. Each moment was sustained for 30 seconds to allow for viscoelastic effects to dissipate (Pelker et al., 1991). The full test was first performed
without preload and then repeated with each preload apparatus, at loads of 200 N and 400 N, for each of the six specimens. The no-load test was carried out again at the end of all the preload tests, to monitor the effect of the protocol on the flexibility of the spine. No-load tests included a counterweight of 1.8kg (average weight of the upper moulding block, torque wheels and fixtures) fixed to the upper vertebra in a manner which preserved all six degrees of freedom of the specimen (Figure A 5-1). Each specimen was therefore subjected to (including before and after no-load tests) ten tests (4 types x 2 magnitudes + 2 no-load) in four directions (flexion, extension and bi-lateral bending) and eight tests (3 types x 2 magnitudes + 2 no-load) in two directions (bi-lateral torsion). The total time to prepare and test each specimen was always less than 20 hours. Exceptions occurred in the testing protocol that prevented data from being collected in certain directions, for three specimens. The omissions occurred only in the initial no-load test and for the left lateral bending and right axial torsion; and did not affect most of the presented data. For clarity, the number of specimens analysed has been included in the results.

A5.3.4. KINEMATIC MEASUREMENT

Four infrared light emitting diodes (IRLEDs), were attached to each PMMA block (Figure A 5-1). The IRLED positions were measured at the end of each load step using an optoelectronic camera system (Optotrak 3020, Northern Digital, Waterloo ON, Canada). Rotations of the upper vertebra with respect to the lower vertebra were expressed as Euler angles (using the ZYX ordering convention). Rotations for each step were defined as the difference between its rotation in the second cycle and the initial (cycle one) balance point. This ensured that the specimen's neutral zone would be included in the rotations reported (Panjabi, 1988).
A5.3.5. Reaction Load Measurement

A six-axis load cell (Advanced Mechanical Technology Inc., MC3A-6-1000, Watertown, MA, USA) was fixed to the bottom of the lower vertebra (Figure A 5-1) and collected continuous three dimensional moment and force data at a rate of 2 Hz. Forces and moments were transformed from the internal load-cell co-ordinate system to a co-ordinate system at the geometric centre of the disc (Figure A 5-1). The moments measured at the disc \( \mathbf{M}_{k,y,z} \) and the moment applied \( \mathbf{M}_a \) were subtracted to give the amplification in applied moment caused by preload application. The additional moment was an artefact of the application method and will be referred to herein as the artefact moment. Measured shear forces will also be reported.

A5.3.6. Statistical Analysis

To evaluate the effect of the test protocol (no-load initial to final), matched pairs t-tests were used. Differences between reaction loads or kinematics were analysed using a two factor repeated measures analysis of variance (ANOVA) with preload type and magnitude as the repeating factors. To evaluate the effect of adding a preload compared to not, type and magnitude data were combined, and a one-way ANOVA was performed with each of the combinations and the final no-load results. Both of these tests were repeated individually for each direction for rotation, moment and force data. Post hoc comparisons were made with the Student-Neuman-Keuls test. Differences between the experimental and calculated artefact loads were evaluated using matched pair t-tests. Linear regression analysis was used to establish the significance of the relationship between the artefact moment and the corresponding rotations, at each level of applied moment. Differences were deemed significant at \( p<0.05 \).

A5.3.7. Mathematical Models

Mathematical models, based on simplified static equilibrium equations, were developed. The production of artefact shear force and moment as a function of rotation
and guide configuration and location were analysed (Figure A 5-3). The equations as presented are applicable to pure flexion and lateral bending rotations but not for torsion rotations. To compare the model and experimental results, the angle $\theta$ was taken as the measured flexion angle and the guide locations (L1 and L2) were directly measured for each specimen.

**A5.4. Results**

There was no statistically significant difference between the initial and final no-load tests, although a trend of increasing rotations was observed. Therefore, all comparisons of load cell data and range of motion (ROM) ratios were made with respect to the more-conservative final no-load case. For brevity, only the results from flexion, extension, right lateral bending and left axial torsion rotations, moments and forces have been presented.

**A5.4.1. Kinematics**

*Flexion:* Average rotations with Type I and Type II were greater than the no-load case at each applied moment step, with a high initial increase observed at the first applied moment step of $M_a = 1.25 \text{ Nm}$ (Figure A 5-4A). The Type I and II ranges of motion (ROM), at 5 Nm, for the 400 N and 200 N preloads tended to increase, by approximately 30 percent and 13 percent respectively (Figure A 5-5A), with respect to no-load and the other preload types. This increase was significantly different from the preload Types III and IV, but not from no-load. For Type I and II, the effect of increasing the preload from 200 N to 400 N within types on the ROM was highly significant ($p=0.004$ for Type I and $p=0.001$ for Type II). ROM for Types III and IV showed only slight variations away from the no-load results.
Extension: All preload types had greater average rotations when compared with no-load, though none of the differences at 5 Nm were significant. (Figure A 5-4B and Figure A 5-5B).

Right Lateral Bending: Ratios of ROM to no-load exhibited a trend amongst types similar to that of flexion, although in this direction the ratios were often less than one (Figure A 5-5C). There was for Type I, however, a high initial increase in average rotation which was maintained across all steps of applied moment (Figure A 5-4C).

Left Torsion: ROM tended to decrease from no-load across all types and loads. The decreases ranged from 22 to 40 percent and were significant in the most extreme cases.

A5.4.2. Moment Data

Flexion, Extension, and Lateral Bending: The average difference between moment measured at the disc centre, for the directions: flexion (+Mx), extension (-Mx) and lateral bending (Mz), and the moment applied (M_a) was relatively high for Type I and Type II, at both load magnitudes (Figure A 5-6A,B,C). As illustrated in Figure A 5-7A, B and C, the induced moments at M_a=5 Nm for Types I and II had increased substantially (0.000 < p < 0.044). Type I, 400 N resulted in the highest increase in moment (40-94%) followed by Type II, 400 N (40-50%) and Types I and II, 200 N (15-40%). This effect was statistically significant with a few exceptions. Types III and IV exhibited no significant change in moment. Interestingly, Type IV induced small oscillations in measured moment (maximum 8.8 %, Figure A 5-6A,B,C). The oscillation was also reflected in the average rotation graphs of Type IV where increases in rotation, for this same region of loads, were also observed to oscillate (Figure A 5-4A,B,C).

Torsion: The effect on moment was very small and consistent for both Type I (a 7.1% and 10.4% increase) and Type II (a 0% and 4.4% decrease). Type III induced an approximately linear decrease in moment as the applied moment was increased. The
moment loss at $M_A=5$ Nm was significant compared to that for the no-load case (a 24% decrease for both 200 N and 400 N of preload).

**A5.4.3. Shear Forces**

*Flexion, Extension, and Lateral Bending:* Except for Type I, shear forces increased with increasing $M_A$. This was most pronounced for Types III and IV (Figure A 5-8A,B,C). With a few exceptions, the shear measured with Type III and IV at 5 Nm was significantly greater than the other types and no-load (Figure A 5-9A,B,C). For all types, there tended to be noticeable increases with increasing preload magnitude.

*Torsion:* In this loading mode, shear forces were of relatively small magnitude (<11 N) and fairly constant.

**A5.4.4. Correlations**

High, positive correlations ($0.785 < r < 0.982; p=0.000$) were found between artefact moment and specimen rotation, for both Type I and II in the directions of flexion-extension and lateral bending (Table A 5-1). Type III and IV in these same directions also produced significant coefficients of correlation, but these were not as strong ($r < 0.720$). In contrast, torsion showed high, negative correlations in Type III ($-0.864 > r > -0.903; p=0.000$) but not in Types I and II. The negative nature of the correlation coefficient indicates that moment is decreasing with increasing axial rotations.

**A5.4.5. Mathematical Models**

The model results are compared to the experimental values in Figure A 5-10. The models were most accurate at predicting artefact moments, where no statistical differences between the experimental and theoretical results occurred (Figure A 5-10B). For shear force prediction, the relative differences between the types were correctly reflected in the theoretical results but they significantly differed from the experimental values for Types I, III and IV (Figure A 5-10A). The largest over or under prediction occurred for Type IV (a 45% over prediction).
A5.5. Discussion

For the first time, this study has established that some common methods of preload application can result in substantial differences between the intended moment and the moment that is actually applied to the specimen during flexibility tests.

It is difficult to define what the desired preload should be. The loads acting on the lumbar spine in vivo are a complex combination of muscle forces and the weight of the body above the segment. It is the complexity of this system that has resulted in the widespread practice of in vitro flexibility testing using isolated moments. This is not believed to represent the in vivo situation but allows repeatable data to be produced and quantitative comparison of surgical techniques and devices. The goal of investigators who wish to add compressive forces to this model is often to produce an "isolated" compressive force to combine with the isolated moments. In this way, the paradigm of "deconstructed" spinal behaviour is preserved as much as possible. To define "isolated" compression, investigators have identified the so-called balance point\(^{16,19}\) and developed mechanical guides and other methods to constrain the preload force vector so that it passes through or close to the balance point irrespective of the specimen rotation.\(^6,7\)

Based on this, it was our view that the goal for a compressive preload device is to result in reaction forces at one end of the spinal segment which were exactly equal to the "pure" axial compression and "pure" moment applied at the other end. Extraneous moments were interpreted as artefacts of the preload application. Extraneous shear forces were also reported although it is unclear if they are artefacts of the preload in the same sense as the extraneous moment.

The artefact moments and the shear forces were considerable in some cases and varied widely depending on preload application method. For Type I, the preload force vector remained vertically oriented (due to gravity) regardless of the amount of rotation of the cranial vertebra. This resulted in an anterior translation of the preload
vector that created a moment arm between the preload vector and the co-ordinate system (Figure A 5-3A). Only in torsion, where the rotation was about the vertical axis, was there no opportunity for the creation of the artefact moment as the test moment was applied. Also, because this preload type incorporated no caudal constraint, virtually no shear forces were developed.

It was somewhat surprising that the differences between Type I and the no-load kinematic responses were not statistically significant. This was likely caused by the relatively high variability, typical of biologic specimens, in the no-load data. The high correlation between kinematic and moment variables supports this hypothesis. Therefore, we believe if either higher moments had been applied or if more specimens had been included in the protocol, Type I would have significantly increased flexion and lateral bending flexibilities as Panjabi et al.12 concluded using this technique. These results confirm the common suspicion that such preloads result in apparent increases in flexibility which actually are caused by the addition of artefact moment. The lack of constraint appears to be advantageous in torsion where this type approaches the “ideal” case where no artefact moment is produced.

The caudal guide incorporated into Types II, III and IV caused the preload vector orientation to rotate as the cranial vertebra rotated. In flexion, extension and lateral bending the preload vector rotation was smallest for Type II, greater for Type III and a maximum (equal to the cranial vertebra rotation) for Type IV. The associated moment arms to the co-ordinate system were thus reduced compared to Type I. The more highly constrained systems (Types III and IV) had the lowest artefact moments but this came at the cost of shear force production in flexion, extension and lateral bending and moment loss in torsion.

For Type IV, the artefact moment tended to oscillate especially in the first few steps of applied moment. It isn’t clear what caused this oscillation but we believe it
indicates physical instability or non-linearity inherent in the lumbar spine. The fact that this behaviour was pronounced for lower values of applied moment suggests it could be related to the neutral zone of the specimen. This is consistent with the established behaviour of the neutral zone, in which relatively large rotations can occur as a result of very low applied moments or forces.\textsuperscript{18}

There were significant reductions or strong trends toward reduced ROM for all directions and types when the artefact moments were close to zero (Type I: torsion, Types III and IV: lateral bending and flexion/extension). These results suggest the existence of physiologic mechanisms that may cause spinal segments to stiffen, for all loading directions, in response to pure axial compression. Possible mechanisms could include facet contact and/or locking\textsuperscript{13} and disc stiffening in response to increased internal pressure.\textsuperscript{7}

The reaction loads were resolved with respect to a co-ordinate system that was fixed to the caudal vertebra. A similar analysis could have been performed with respect to a co-ordinate system that rotated with the cranial vertebra. The results with respect to artefact moment would be the same for both co-ordinate systems. In shear, Type I would produce the largest shear forces and Type IV would produce no shear with respect to the cranial vertebra. It is difficult to make an overwhelming case for selecting one of the co-ordinate system approaches over the other. Our choice of co-ordinate system allowed us to view the FSU as consisting of an "input" vertebra (the cranial vertebra) to which we applied the isolated moment and compressive preload and an "output" vertebra (caudal) where we measured the resulting loads.

The results presented are specific to single FSUs. The extension of these results to testing segments of longer lengths (i.e. two or more FSUs) is naturally of great interest since this is also a common testing configuration. The application of Type I or II preloads is straightforward but in both cases significantly larger artefact forces and moments are
likely to occur owing to the larger total rotations occurring over longer spine segments. The adaptation of Type III or IV preloads to longer segments is technically challenging. Based on the study by Patwardhan et al., small eye-hooks could be screwed to the lateral aspects of each vertebral body which could act as guides for the preload application wire. Our results for Type III suggest such a design could produce considerable artefact moments in torsion loading, which would act to reduce the effective moment applied to the specimens and be proportional to the rotation angle at each level. The analytical techniques used in this investigation could easily be adapted to evaluate preload application techniques over longer segments.

The equilibrium equations presented in Figure A 5-3 imply several simplifying assumptions. The analysis presented is nevertheless adequate for quantitative comparison or parametric study of the effects of altering the guide location, guide configuration or specimen rotation on artefact moments and forces.

The results of this study provide the basic data necessary to increase the fidelity between in vitro spine tests and the loading situation in vivo. The preload magnitude and application method have both been shown to affect kinematic response and the effective loads that the specimen experiences. This study emphasises the necessity to report precisely how axial preload is applied in spinal flexibility tests. Failure to do this may make it impossible to replicate results from individual studies and render results from seemingly similar studies inappropriate to compare.

There was a clear mechanical "trade off" evident in the results, whereby high artefact moments present for the relatively unconstrained preload application methods like Type I could be reduced or eliminated by adding appropriate guides but only at the cost of shear force production. In vivo spinal loads are not composed of isolated moments and forces. When this loading situation becomes clearly understood it may be that the so-called artefact moments and the shear forces measured here do in fact
reflect those occurring physiologically. If the goal of an in vitro flexibility study is to superimpose isolated compressive forces with isolated moments, our results suggest a Type III or IV preload application method for flexion, extension and lateral bending coupled with a relatively unconstrained method like Type I or II for torsion will most accurately produce the desired loading patterns.

Acknowledgements: The authors thank Cara Inglis for assistance preparing the figures.

A5.6. References


The PMA blocks rotate with the cervical vertebrae. The infra-red-light-emitting diodes can be seen mounted to
indicate the co-ordinate system with two vertebrae and hence did not
fix to the load cell fixed below the caudal vertebrae and one of the two counterweights are
left axial rotation without preload. The orientation of the co-ordinate system, the location

Figure 5.1: Three-dimensional moment application apparatus set to apply moment in

A5.7 Figures
Figure A 5-2: The four types of preload application configured for flexion/extension testing: A) Type I: a weight was hung from a point above the cranial vertebra using steel preload cables. The anterior/posterior location of the preload wire was adjusted to find the balance point. B) Type II: a guide was added at the base of the load cell C) Type III: the caudal guide and cranial fixation point were extended to the upper and lower endplates. D) Type IV: the preload vector was maintained parallel to the cranial endplate using the lined plate and adjustable lower guides shown.
Figure A 5-3: Free body diagrams and relevant static equilibrium equations for each of the preload application methods: A) Type I, B) Type II and Type III, and C) Type IV. In Figure B), the guides are illustrated at an intermediate position between the extremes represented by types II and III. The sagittal plane shear forces and artefact moments, which occur as a result of a flexion rotation, $\theta$ were calculated. Pure rotation of the specimen about the centre of the disc (point O) was assumed.
Figure A 5-4: The average specimen rotation, as a function of applied moment, $M_A$, for A) flexion, B) extension and C) lateral bending moment application. The responses with no-preload and with 400 N of preload applied using each of the four application methods are presented.
Figure A 5-5: The average ratio, at $M_A = 5 \text{ Nm}$, of the ROM for each preload type and magnitude to the ROM for the final no-load case. Ratios are presented for A) flexion, B) extension and C) lateral bending moment application. Error bars represent one standard deviation. Significant differences ($P<0.05$) are indicated at the top of each bar. Significance to no-load is indicated by an $a$. 
Figure A 5-6: The average artefact moment induced, $M_{x,y,z} - M_A$, as a function of applied moment, $M_A$, for A) flexion, B) extension and C) lateral bending moment application. $M_{x,y,z}$ = net moment sensed and $M_A$ = applied moment. The responses with no-preload and with 400 N of preload applied using each of the four application methods are presented.
Figure A 5-7: The average net moments sensed for each preload type and magnitude, at $M_A = 5$ Nm. Data are presented for A) flexion, B) extension and C) lateral bending moment application. The error bars represent one standard deviation. Significant differences ($P<0.05$) are indicated at the top of each bar. The artefact moment $M_{x,y,z} - M_A$ is represented by the portion which exceeds the horizontal line at $M_A = 5$ Nm.
Figure A 5-8: The average shear force induced, $F_{XZ}$, as a function of applied moment, $M_A$, for A) flexion, B) extension and C) lateral bending moment application. Average anterior-posterior shear has been plotted for flexion and extension, and medial-lateral shear has been plotted for lateral bending. The responses with no-preload and with 400 N of preload applied using each of the four application methods are presented.
Figure A 5-9: Average shear forces sensed for each preload type and magnitude, at $M_a = 5$ Nm. Data are presented for A) flexion, B) extension and C) lateral bending moment application. The error bars represent one standard deviation. Significant differences ($P<0.05$) are indicated at the top of each bar.
Figure A 5-10: Comparison between the measured artefact shear forces, A and artefact moments, B and the corresponding values calculated using the equations in Figure A5-3. The illustrated cases are for flexion loading with $M_A = 5$ Nm and an applied preload of 400 N. An asterisk indicates significant differences between the theoretical and experimental results at $P<0.05$. 
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<th>Mz (n=44)</th>
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<td>400 N</td>
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<td>0.112 (NS)</td>
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<tr>
<td>Type III</td>
<td>200 N</td>
<td>0.451 (0.001)</td>
<td>0.358 (0.010)</td>
<td>-0.903 (0.000)</td>
</tr>
<tr>
<td></td>
<td>400 N</td>
<td><strong>0.720 (0.000)</strong></td>
<td><strong>0.520 (0.000)</strong></td>
<td>-0.864 (0.000)</td>
</tr>
<tr>
<td>Type IV</td>
<td>200 N</td>
<td><strong>0.503 (0.000)</strong></td>
<td><strong>0.608 (0.000)</strong></td>
<td></td>
</tr>
<tr>
<td></td>
<td>400 N</td>
<td><strong>0.639 (0.000)</strong></td>
<td><strong>0.569 (0.000)</strong></td>
<td></td>
</tr>
</tbody>
</table>

Table A 5-1: Correlation coefficients for the relationship between kinematic response (in degrees, vertical axis) and the artefact moment ($M_{xyz} - M_A$ Nm, horizontal axis). Coefficients in bold represent high correlations ($r > 0.5$). Statistical significances are presented in brackets.