Spinal modelling to investigate postural loading and stability

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ABSTRACT

Numerous mathematical models have been developed to investigate the high incidence of low back pain associated with lifting activities. These mainly consider the muscle forces required to support the spine, and few have considered the additional role of curvature.

One previous model which represented the spine as an arch (Aspden 1987) indicated the curvature to have a significant effect on both loading and stability of the spine. However this model included collective loading patterns for body weight and muscle forces, and only partial representation of the spine. On the basis that the level of anatomic detail of a model affects the accuracy of its predictions (McGill and Norman, 1987), this thesis describes the development of a model which provides greater detail for investigating spinal stability in the sagittal plane.

The curvature of the whole spine, a distributed loading pattern for body weight, and the activity of individual spinal muscle groups have been considered. Comparison with the previous arch model has shown these to be necessary features for determining the loading and stability associated with a given posture. In particular, application of individual muscle forces provide greater control of stability at each vertebral level. By considering the force requirements of the individual muscle groups and the consequent loads at each intervertebral joint, possible areas of tissue over load can be identified.
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# TABLE OF CONTENTS

## CHAPTER ONE:

1. INTRODUCTION ................................................. 1

## CHAPTER TWO:

2. THE NEED FOR MATHEMATICAL MODELS OF THE HUMAN SPINE ................................................. 6
   2.1. The Complexity Of The Human Spine ............... 6
   2.2. Abnormalities Of The Human Spine ............... 11
   2.3. The Problem Of Low Back Pain .................... 14
   2.4. Comparison Of Spinal Research Techniques ....... 18
       2.4.1. Requirements Of Spinal Research ............ 18
       2.4.2. Comparison Of Experimental And Mathematical Modelling Research Techniques .......... 20
   2.5. A Critical Review Of Existing Spinal Models .... 26
       2.5.1. Whole Spine Models .......................... 30
       2.5.2. Partial Spine Models: Head And Neck Models ........... 37
       2.5.3. Partial Spine Models: Thoraco-Lumbar Region .......... 47
       2.5.4. Partial Spine Models: Lumbar Region Models ........ 61
       2.5.4.1. Single Muscle Force Models ................. 62
       2.5.4.2. Anatomical Representation Of The Muscle Forces ........... 67
       2.5.4.3. Three Dimensional Models ................. 68
       2.5.4.3.1. Optimisation Assisted Models .......... 70
       2.5.4.3.2. Electromyography Assisted Models .......... 72
       2.5.4.4. Ligaments And Intra Abdominal Pressure ........... 77
       2.5.4.5. Multi-Segmental Models ................. 81
2.5.5. Unit Models 94
2.5.5.1. Cervical Unit Models. 94
2.5.5.2. Thoracic Unit Models 97
2.5.5.3. Lumbar Unit Models 98
2.5.5.3.1. Vertebrae 99
2.5.5.3.2. Disc_Body Unit Models 102
2.5.5.3.3. Motion Segments 107
2.5.6. Discussion 114
2.6. The Purpose Of A New Mathematical Model 125

CHAPTER THREE:

3. DEVELOPMENT OF A NEW MATHEMATICAL MODEL OF THE HUMAN SPINE 129

3.1. Equilibrium And Stability Of An Arch Structure 130
3.1.1 Equilibrium 130
3.1.2 Stability 132
3.2. Aspden's Application Of Arch Theory To The Spine 135
3.2.1 Conditions of Arch Theory 135
3.2.2 Postures Investigated 137
3.3. Features Of The New Model 139
3.3.1. Whole Spine 139
3.3.1.1. Curvature 140
3.3.1.2. End Supports 141
3.3.2. Distributed Forces Due To Body Weight 143
3.3.3. Distributed Forces Due To Muscles And Ligaments 143
3.3.4. Variation In Posture 145
3.3.5. Intervertebral Joint Reactions 146
3.3.6. Two Dimensional Representation 146
3.3.7. Model Applications 147
3.4. Aims And Objectives 148
3.5. Human Spine Modelling Using Computer Based Modelling Tools 149
3.5.1. 'Adams': A Computer Based Modelling Application 151
3.5.2. Mathematical Representation Of The Spinal System Using Adams 152
3.5.3. Model Input 153
3.5.3.1. Mathematical Representation Of The Spinal Configuration 154
3.5.3.2. Geometrical Representation Of The Vertebrae 157
3.5.3.3. Boundary Conditions For The Model 158
3.5.3.3.1. Arch Limits 158
3.5.3.3.2. Forces At The Upper Arch Boundary 159
3.5.3.3.3. Forces At The Lower Arch Boundary 161
3.5.3.4. Definition Of The Applied Spinal Forces 163
3.5.3.4.1. Body Weight Forces 163
3.5.3.4.2. Muscle Forces 163
3.5.4. Model Calculations 167
3.5.4.1. Calculation Of The End Support Reactions For An Arch Structure Using Adams Command Language 167
3.5.4.1.1. Perpendicular End Support Reaction Components R1 And R2 168
3.5.4.2. Calculation Of The Thrustline Co-ordinates For An Arch Structure Using Adams Command Language 173
3.5.4.2.1. Sorting Procedure 176
3.5.4.2.2. Equal Locations 178
3.5.4.3. Graphical Display Of The Predicted Thrustline 179
3.5.4.4. Calculation Of The Joint Reactions At Each Intervertebral Level Using Adams Command Language 179
3.5.4.5. Output Of Joint Reactions, Thrustline Co-ordinates And Muscle Forces 181
3.5.5. Summary 182
3.6. Derivation Of Input Data For The Model

3.6.1. Simulation Of Vertebral Configuration For Various Postures

3.6.1.1. Determination Of Vertebral And Disc Dimensions And Orientations

3.6.1.2. Configuration In The Erect Posture

3.6.1.3. Configuration In The Flexed Postures

3.6.2. A Distributed Loading Pattern For Body Weight Along The Spine And The Addition Of External Loads

3.6.2.1. Derivation Of A Force Distribution Pattern For Body Weight

3.6.2.2. Additional External Loads Carried In The Arms

3.6.3. Derivation Of A Distributed Pattern Of Muscle Forces Acting Along The Spine

3.6.3.1. Direct And Indirect Muscle Forces Acting On The Spine

3.6.3.1.1. Direct Forces

3.6.3.1.2. Indirect Forces

3.6.3.2. Selection Of Muscle Groups For Investigating Spinal Stability

3.6.3.3. Points Of Application Of Direct Muscle Forces

3.6.3.4. Points Of Application Of Indirect Forces

3.6.3.5. Representation Of The Points Of Muscle Attachment

3.6.3.5.1. Muscle Attachment To The Vertebrae

3.6.3.5.2. Muscle Attachments To The Pelvis And Sacrum

3.6.3.5.3. Muscle Attachments To The Head

3.6.3.5.3.1. Atlas And Axis Vertebra

3.6.3.5.3.2. Attachments To The Skull

3.6.3.5.3.2.1. X Co-ordinates In The Sagittal Plane

3.6.3.5.3.2.2. Y Co-ordinates In The Sagittal Plane

3.6.3.5.4. Muscle Attachments To The Ribcage

3.6.3.6. Lines Of Action Of The Muscle Forces

3.6.3.7. Determination Of The Magnitude Of The Muscle Forces Acting On The Spine
3.6.3.7.1. Data Available In The Literature For The Cross Sectional Areas Of The Spinal Muscles

3.6.3.7.1.1. Lumbar Muscles

3.6.3.7.1.2. Cervical Muscles

3.6.3.7.1.2.1. Cadaveric Studies

3.6.3.7.1.2.2. Studies On Living Subjects

3.6.3.7.1.2.3. Additional Study

3.6.3.7.2. Selection Of Nominal Cross Sectional Area Values

3.6.3.7.2.1. Unavailable Data

3.6.3.7.2.2. Intersegmental Muscles

3.6.3.7.2.3. Lumbar Muscle Forces

3.6.3.7.2.4. Abdominal Muscles

3.6.3.7.3. Force Magnitude

3.6.3.7.4. Force Distribution

3.6.4. Ligament Forces In The Sagittal Plane

3.6.4.1. Estimation Of Ligament Cross Sectional Area

3.6.4.2. Estimation Of Ligament Stress.

3.6.4.3. Estimation Of Ligament Strain With Flexion.

3.6.4.4. Application Of Theoretical Ligament Forces To The Model Developed In ADAMS

3.6.4.5. Alternative Points Of Attachment

3.6.5. Forces Due To Intra Abdominal Pressure
CHAPTER FOUR:

4. INVESTIGATING STABILITY AND LOADING OF THE SPINE
   USING THE DEVELOPED COMPUTER-BASED MODEL

4.1. The Action Of Body Weight On The Spine Represented As An Arch Structure
   4.1.1. Model Predictions
   4.1.2. Explanation Of The Predicted Thrust Lines
      4.1.2.1. Force Direction
      4.1.2.2. Changes In Point Of Force Application With Posture
      4.1.2.3. Generated Moments
         4.1.2.3.1. Erect Posture
         4.1.2.3.2. Flexed Postures
      4.1.2.4. Effects Of Forces And Moments On Predicted Thrustline
         4.1.2.4.1. Erect Posture
         4.1.2.4.2. Flexed Postures
   4.2. The Action Of External Loads
      4.2.1. Model Predictions
      4.2.2. Explanation Of The Effects Of External Loads
         4.2.2.1. Force Direction
         4.2.2.2. Point Of Force Application
         4.2.2.3. Generated Moments
            4.2.2.3.1. Erect Posture
            4.2.2.3.2. Flexed Postures
         4.2.2.4. Effects Of Forces And Moments On The Predicted Thrustline
      4.2.3. Discussion
4.3. The Effects Of The Action Of Each Muscle Group On The Predicted Thrustline For The Spine As An Arch

4.3.1. Model Predictions For The Erect Posture

4.3.2. Analysis Of The Muscle Group Behaviour On The Predicted Thrustline For The Spine As An Arch In The Erect Posture

4.3.2.1. Muscle Groups Attaching To The Sacrum Or Pelvis

4.3.2.1.1. Lumbar Muscles

4.3.2.1.2. Abdominal Muscle Groups

4.3.2.2. Effects on R2

4.3.2.3. Muscle Categorisation

4.3.2.4. Muscle Groups Acting Above The Sacrum And Pelvis

4.3.2.4.1. Force Application

4.3.2.4.2. Muscle Categorisation

4.3.2.4.2.1. Cervical Muscles Between The Vertebrae And Head

4.3.2.4.2.2. Cervical Muscles Between Vertebrae

4.3.2.4.2.3. Thoracic Region Muscles Between Vertebrae

4.3.2.4.2.4. Intersegmental Muscles

4.3.2.4.2.5. Muscle Groups Attaching Between The Vertebrae And The Ribcage

4.3.3. Model Predictions For Muscle Group Behaviour With Flexion

4.3.3.1. Relative Movement Between Categories.

4.3.4. Analysis Of The Effects Of Flexion On The Action Of Each Muscle Group On The Predicted Thrustline

4.3.4.1. Lumbar Muscles

4.3.4.2. Abdominal Muscles

4.3.4.3. Cervical Muscles

4.3.4.4. Intersegmental Muscles

4.3.4.5. Rib Cage Muscles

4.3.4.6. Relative Movement Between Cervical And Lumbar Muscle Groups

4.3.5. Discussion
4.4. The Effects Of Ligaments On Stability Of The Spine 316
4.4.1. Model Predictions 316
4.4.2. Discussion 318

4.5. Sensitivity Analysis 321
4.5.1. Effects Of Curvature On Muscle Force Direction 322
4.5.1.1. Comparison Of The Predicted Thrustlines For Muscles With Straight Or 'Curved' Force Representation 322
4.5.1.2. Discussion 325
4.5.1.2.1. Thoracic Iliocostalis 325
4.5.1.2.2. Longissimus Thoracis 327
4.5.1.2.3. Spinalis Thoracis, Splenius Cervicis And Longus Coli 330
4.5.1.3. Summary 331
4.5.2. Variation In Force Distribution 332
4.5.2.1. Inter-Segmental Muscles 332
4.5.2.2. Multi-Segmental Muscles 333
4.5.2.3. Model Predictions 334
4.5.2.4. Discussion 337
4.5.3. The Effects Of Force Magnitude On The Predicted Thrustline For The Intersegmental Muscles 340
4.5.3.1. Model Predictions 341
4.5.3.1.1. Erect Posture 341
4.5.3.1.2. Flexed Postures 342
4.5.3.2. Discussion 344

4.6. Muscle Groups Which Are Important For Stability Of The Spine 345
4.6.1. Matching Muscle And Ligament Behaviour With The Force Requirements In Each Posture 345
4.6.2. Muscle Groups Required To Stabilise The Spine When Supporting The Weight Of The Upper Body

4.6.2.1. Erect Posture

4.6.2.2. Flex1 Posture

4.6.2.3. Flex2 Posture

4.6.2.4. Flex3 Posture

4.6.2.5. Discussion

4.6.3. Muscle Groups Required To Stabilise The Spine When Supporting The Weight Of The Upper Body And An External Load Of 900N

4.6.3.1. Erect Posture

4.6.3.2. Flex1 Posture

4.6.3.3. Flex2 And Flex3 Posture

4.6.3.4. Extended Study For The Flex2 And Flex3 Postures

4.6.3.5. Discussion

CHAPTER FIVE:

5. MODEL EVALUATION AND VALIDATION

5.1. Evaluation Of The Model Features

5.1.1. Effects Of Distributed Forces Due To Body Weight On Thrustline Curve

5.1.1.1. Conclusions

5.1.2. The Effects Of Distributed Body Weight, Muscle And Ligament Forces On Predicted Thrustlines

5.1.2.1. Direction Of The Thrustline Path

5.1.2.2. Indication Of Stability

5.1.2.3. Requirements For Muscle Force

5.1.2.4. Amount Of Muscle Force Required

5.1.2.5. Summary
Chapter 5

5.1.3. The Effects Of Distributed Body Weight Forces On Predicted Joint Reactions

5.1.3.1. Observations Common To Both Loading Patterns

5.1.3.1.1. Intervertebral Compression

5.1.3.1.2. Intervertebral Shear

5.1.3.2. Distinction Between Collective And Distributed Loading Patterns

5.1.4. The Effects Of Distributed Body Weight and Muscle And Ligament Forces On Predicted Joint Reactions

5.1.4.1. Compression

5.1.4.1.1. Erect Posture

5.1.4.1.2. Flex2 Posture And Body Weight

5.1.4.1.3. Flex3 Posture And Body Weight

5.1.4.1.4. Flex1 Posture And External Load

5.1.4.1.5. Flex3 Posture And External Load

5.1.4.2. Shear

5.1.4.3. Discussion

5.1.4.3.1. Compression

5.1.4.3.2. Shear

5.1.5. Representation Of The Whole Spine

5.1.6. Method Of Investigation

5.1.7. Model Performance

5.1.7.1. Program Run Time

5.1.7.2. Alteration Of Parameter Values

5.2. Model Validation

5.2.1. Joint Reaction Predictions And Intra Discal Pressure Measurements

5.2.1.1. Erect Posture

5.2.1.2. Flexed Postures

5.2.1.3. Discussion

5.2.2. Muscle Roles And Recruitment Patterns

5.2.3. The Role Of The Ligaments
5.2.4. Model Predictions And Existing Concepts 429
  5.2.4.1. Muscle Coactivity 429
  5.2.4.2. Flexion - Relaxation Phenomenon 432
  5.2.4.3. Intra Abdominal Pressure 434
5.3. Model Limitations And Developments 435
  5.3.1. Whole Spine Configuration 435
  5.3.2. Application Of The External Load 436
  5.3.3. Muscle And Ligament Force Distribution Patterns 437
  5.3.4. Intra Abdominal Pressure 438
  5.3.5 Curved Muscle Representation 439
  5.3.6. Dynamic Investigations 440
  5.4. Application Of The Model To Future Low Back Pain Investigations 441

CHAPTER SIX:

6. CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE RESEARCH 444
  6.1 Basic Conclusions 444
  6.2 Recommendations for Future Research 450

7. REFERENCES 452
  7.1 Bibliography 475
8. APPENDIX

8.1. Validation Of The Relative Angles Used For Vertebral Configurations

8.2. Representation Of The Points Of Muscle Attachment To The Vertebrae

8.2.1. Calculation Of Location Of Spinous Process Relative To Vertebral Origin

8.2.2. Calculation Of Location Of Transverse Process Relative To Vertebral Origin

8.2.3. Calculation Of Location Of Superior Articular Process In The Cervical Region Relative To Vertebral Origin

8.2.4. Calculation Of Location Of Mamillary Processes In The Lumbar Region Relative To Vertebral Origin

8.2.5. Calculation of Location of Vertebral Lamina in the Thoracic Region Relative to Vertebral Origin

8.3. Calculation Of Co-ordinates Of Points Of Attachment Of Muscle Groups To The Skull

8.3.1. X Co-ordinates

8.3.2. Y Co-ordinates

8.4. Thrustline Categorisation

8.4.1. Erect Posture

8.4.2. Flex1 Posture

8.4.3. Flex2 Posture

8.4.4. Flex3 Posture

8.5. Illustrations Of The Lines Of Action Of Muscle Groups Referred To In The Text

8.5.1. Straight And Curved Representations Of The Lines Of Action For The Muscle Groups Identified To Be Affected By Curvature
CHAPTER ONE:

1. INTRODUCTION

The focus of this work was to develop a mathematical model of the spine which would allow the forces in the muscles and intervertebral joints to be predicted for a given posture. The motivation for this work has arisen from the need of many who suffer from back pain, the precise causes of which are still not yet fully understood.

Classification of the existing models in the literature reveals whole, partial and unit spine models. Numerous partial models of the lumbar spine investigate the underlying causes of low back pain associated with lifting activities. The model described in this thesis accompanies the low back pain investigations, but includes a description of the whole spine so that the influence of posture can be considered.

A number of low back pain models investigating lifting activities assume the lever mechanism to be an appropriate representation for the spine. Equilibrium of forces and moments about the intervertebral joints are considered. The body weight and load lifted generate a flexion moment which pulls the trunk forward. The amount of force necessary to generate an equivalent extensor moment, and the consequent joint reaction forces are then determined. It is assumed that the greatest forces and moments are generated at the lower lumbar intervertebral joints, at which a high incidence of low back pain is reported.
Early models were restricted to a single muscle force acting at a single intervertebral level. However, the predictions for intervertebral joint compression exceeded the known failure values of the vertebrae for lifts which were performed successfully by living subjects. Realisation that the simple representation of the muscle forces overestimated spinal loading, has led to an increase in the level of anatomic detail for the muscles, and predictions for the joint reaction forces were reduced.

The model predictions for joint compression have been used to define guidelines for lifting strategies and load limits in industry. However, despite this, a high incidence of low back pain has still been reported.

Mechanically, both equilibrium and stability are necessary in order to prevent collapse of a mechanical system. More recently, possible causes of low back pain have therefore been attributed to instability of the lumbar spine. Multisegmental models developed to investigate lumbar stability are based upon Euler buckling theory. The amount of muscle force required to provide sufficient stiffness to the spine to resist an applied load is determined for a given configuration. These models have consequently attributed roles to the smaller spine muscles for which the lever model could not account. However, these models have mainly been restricted to consideration of the lumbar spine.

While research is continuing in this direction, an alternative mechanism which provides a significant role for the curvature of the spine has been overlooked. Aspden (1987) proposed that representation of the spine as an arch showed significant reductions in the amount of muscle force required and the consequent loads on the intervertebral joints. The curvature of the spine may therefore have a significant role in determining joint and muscle loading.
This model also provides a graphical representation of the state of stability of the spine. The thrustline for an arch indicates the path of forces through the structure. Transmission of these forces through the vertebral bodies indicates the spine is stable.

However, this model included collective loading patterns for body weight and muscle forces, and only partial representation of the spine. On the basis that the level of anatomic detail of a model affected the accuracy of the lever model predictions (McGill and Norman, 1987), the predictions of the arch model may be affected. Development of this model is therefore necessary to establish the role of the curvature of the spine in determining loading and stability.

The work described in this thesis is unique in its approach to investigating spinal loading in flexion activities. The model developed pursues the concept of arch theory, but includes a more detailed loading pattern for body weight, and individual muscle groups. Based on the understanding that posture can influence the loading on the spine, the model includes the curvature of the whole spine. The model therefore provides a link between posture and spinal loading.

In comparison with existing lever models, the stability of the spine is considered. The model also elaborates upon existing stability studies of the lumbar region by considering the whole spine. Finally, as a development of the existing arch model, the role of the individual muscle forces and the consequent loading at each intervertebral joint can be considered.

The developments of this model, the investigations performed, an evaluation of the model performance and predictions, are described in more depth in this thesis. The following text briefly describes the material contained in each chapter of this thesis.
Chapter Two
Chapter two describes the complexity of the human spine, and illustrates the difficulty in identifying causes of low back pain. Among several spinal disorders, low back pain is identified to be a distinct problem which requires research. The preference for mathematical modelling techniques is illustrated through a comparison with experimental methods. This is coupled with a review of a number of models, illustrating their contribution to a vast range of applications. Comparisons with existing models identifies significant features which may be beneficial to low back pain investigations. The development of a model which explores the role of the curvature of the spine and its influence on muscle and joint loading is proposed.

Chapter Three
The development of the model is discussed in Chapter three. First the concept of arch theory, and its application to the spine proposed by Aspden (1987) is examined. Proposals for the application of this theory to accommodate greater anatomic detail of the spine including the curvature of the whole spine, and distributed force patterns for the body weight and muscle forces are presented. Available data in the literature regarding spinal configurations, body weight distribution and the characteristics of each muscle group is then used to determine input data for the model.

The model was developed using the computer based modelling tool 'ADAMS', and run on a SUN_SPARC_20 machine. The application provided a graphical interface which allowed immediate visible feedback of results. Algorithms written to define the relation between the spinal curvature and applied forces within the computer based modelling application 'ADAMS' are then discussed.
Chapter Four
Using the model developed in Chapter three, the stability of the spine is first considered due to the action of body weight only, and with the addition of an external load. The contribution of each individual muscle and ligament group towards the stability of the spine when supporting these loads are then considered. The force requirements for stability for each posture are matched by combinations of muscle groups. The capacity of the muscles to support the weight of the upper body and an additional load in each posture is thus investigated.

Chapter Five
This chapter evaluates the model predictions, by comparison of the intervertebral joint reactions, the muscle force required, and the state of stability with a collective loading pattern used in the previous model (Aspden, 1987). Comparison of the model predictions with reported EMG recordings of muscle recruitment, and intervertebral disc pressure measurements is also made.

Chapter Six:
The concluding chapter of the thesis reviews the findings of this work in relation to existing concepts and presents ideas towards the use of the model in future back pain investigations.
CHAPTER TWO:

2. THE NEED FOR MATHEMATICAL MODELS OF THE HUMAN SPINE

The complexity of the spine makes it difficult to understand its normal and abnormal behaviour. Research is necessary to help prevent or correct disorders. In particular, there is a high prevalence of low back pain in many Westernised societies. Research methods include experimental techniques and mathematical modelling techniques. The latter is a more favourable research method because it avoids the risk of injury to living subjects, and involves less cost than with cadaver experiments. With advances in computer techniques, a number of mathematical models have been developed to investigate the behaviour of the spine. Many of these models have clinical applications. A review of these models helps to identify areas in which models have been beneficial, and areas which could benefit further. Based upon the findings of this review, the requirements for a new model to investigate the causes of low back pain are identified. This chapter follows these issues in greater detail.

2.1. THE COMPLEXITY OF THE HUMAN SPINE

The human vertebral column is one of the most important skeletal structures in the human body. When functioning normally, it transmits loads borne by the upper extremities to the lower limbs, provides mobility to the head and trunk, and protects the spinal cord. However in order to serve these functions, the spine is also the most complex musculoskeletal system.
The vertebrae are irregularly shaped bony elements which constitute the skeletal portion of the spine. Twenty two moveable vertebrae occupy the cervical, thoracic and lumbar regions, and have common geometrical features, allowing them to serve similar functions. Each has an anterior and a posterior region. The vertebral body, which is approximately cylindrical in shape, constitutes the anterior part and has a large cross sectional area to withstand the applied loads.

Posterior elements of the vertebra consist of facets, pedicles, lamina and various processes for muscle and ligament attachment. Together with the posterior surface of the vertebral body, the pedicles and lamina form the neural canal, which encloses and protects the spinal cord. Pairs of facets provide surfaces lined with cartilage for articulation between two adjacent vertebrae. The orientation of these surfaces controls the amount of movement possible between the vertebrae. However, movement cannot be controlled until their surfaces come into contact.

Figure 2-1 shows the spinal column in the upright posture. The constitutive vertebrae are labelled C1-C7 in the cervical region, T1-T12 in the thoracic region, and L1-L5 in the lumbar region. This notation is used throughout the thesis.
Figure 2-1: Illustration of the twenty four moveable vertebrae constituting the spinal column. (C=cervical, T=thoracic, L=lumbar.)

Force transmission between the vertebrae can occur through two pathways. Between the posterior part of the vertebrae, loads can be transmitted between the facets when contact occurs. Alternatively forces can be transmitted between the anterior part of the vertebrae through interconnecting discs, firmly attached to the bony end plates of the vertebral bodies. Under axial compression, the disc is reported to be the primary load-bearing structure (Belytschko et al., 1974). However in activities which involve lateral bending, extension or twisting, the articular facets also play a significant role.
The discs have a complex substructure. The inner region is occupied by a fluid filled sac known as the nucleus pulposus, which consists of approximately 80% water in a normal (non degenerated) state (Nachemson, 1987). Enclosing the nucleus are collagen fibres arranged in 8-12 circumferential layers constituting the annulus fibrosus. Within each layer the fibres are oriented approximately parallel relative to each other, but between layers are oriented at approximately 30° in alternating directions. Throughout the annulus, the fibres are embedded in an amorphous ground substance. The combination of fibres in alternating directions, (and ground substance) reinforces the structure of the annulus. Towards the outer circumferential layers there is a greater percentage of collagen fibres thereby allowing it to resist greater loads in this region. The disc also demonstrates poroelastic properties, with diurnal fluid flow into and out of the disc for the transfer of nutrients and waste products from the adjacent vertebral surfaces.

The multi-layered architecture of the ligaments and muscles results in a complex system of forces generated along the entire length of the spine. The distinguishing feature between the muscle and ligaments is their composition. Muscles contain active contractile elements which allow them to generate force and initiate movement of the spine. The level of force is monitored by neuro-receptors, and the muscle recruitment pattern adjusted to provide the amount of force required. In contrast the ligaments are composed mainly of collagen tissue which relies upon stretching in order to generate tensile forces. When the relative movement of the vertebrae to which the ligament is attached becomes excessive, the tensile force generated within the ligament serves to resist this movement. The ligaments therefore serve to restrict the movement of the spine and prevent excessive motion which may be likely to cause damage to tissues. Passive force is also generated within the muscle tendons upon stretching, allowing the spine to control movement with minimal energy expenditure.
The material and geometric properties of these elements vary with the applied load. The collagen fibres of both the annulus and ligaments, and the articular cartilage of the facets exhibit a non linear stiffening response as the applied load is increased. More rapid loading results in a reduction in the force actively produced by the muscles (Astrand and Rodaal, 1970) but increases the stiffness of the ligaments due to their viscoelastic properties (Hukins et al., 1990), and also increases vertebral strength (Aspden, 1992). Repetitive loading of the ligaments allows them to store energy, while muscle contractile elements can fatigue.

Regional variation in material and geometric properties of the spinal components also occurs. At the upper end of the spine the atlas and axis vertebrae have large facet surfaces for articulation with the occipital condyles of the skull, and at the lower end of the spine articulation of the sacrum with the pelvis allows for transfer of loads from the spine to the lower limbs. The attachment of the ribcage in the thoracic region also restricts movement in this area. Descending the spine there is a general increase in the cross sectional area of the vertebral bodies. This reflects their ability to withstand the cumulative weight of the trunk. Variation in the disc properties and the orientation of the facets also affects the range of movement possible for each region.
The complexity of the spine is thus evident through the considerable variation in material and geometric properties of its components. The range of postures which the spine can adopt results in complex loading, which requires interaction of these components for complete support. The direction of the applied load is influenced by curvature, so that the interactive role of the disc and facets may therefore be different at each level. Changes in loading conditions at the upper levels of the spine has a direct influence upon the loads transmitted to the lower levels of the spine. Also, forces generated at lower levels by muscle elements can equally affect the upper levels through their multi-layered attachments. Interaction of components at each level, and between levels results in a highly complex system.

Abnormality in one component may be compensated by an increase in the roles of other elements, either at that level, or at different levels. The spine may therefore adapt to overcome a damaged component without complete failure of the system. However, prolonged abnormality may eventually cause an overload at other levels and result in inhibited functioning of the overall system.

2.2. ABNORMALITIES OF THE HUMAN SPINE

A visible abnormality of the human spine is scoliosis. Unlike a normal spine which forms a straight column in the frontal plane, a scoliotic spine shows curvature in both the sagittal and lateral planes. The precise causal mechanism for the development of this disorder is not clear, although it may be linked with muscle imbalance, coupled motion of the vertebrae, and asymmetric vertebral development. It results in pain and a reduced load bearing capacity.
Other abnormalities are not externally visible but can be detected by X-ray procedures, CT (Computer Tomography) scans or MRI (Magnetic Resonance Imaging). Vertebral fractures can occur when the applied loads are excessive, for example through impact in pilot ejection or vehicle collision. The likelihood of vertebral fracture is also increased with the incidence of osteoporosis and metastases (spinal tumours).

Wear and tear of the synovial joints is generally described as osteoarthritis. Also with ageing, a build up of bone around the edge of the disc and facet joints can occur, known as spondylosis. The excess may reduce the dimensions of the spinal canal (spinal stenosis) and may impinge on the spinal nerves causing pain, cramp and even reduced blood supply to the lower limbs.

Due to a break in the neural arch of the vertebra, or due to wear and tear, excess movement of the vertebrae may occur. A superior vertebra may slip forwards (spondylolisthesis) or backwards (retrolisthesis) over the inferior vertebra, causing damage to the ligaments, or stretching the nerve roots. Significant pain is caused, especially upon movement.

Other more serious conditions include rheumatoid arthritis and ankylosing spondylosis, involving inflammation of the spinal joints. These involve severe pain and considerable loss of movement.

Accompanying any disorder of the spine is a change in load bearing capacity and range of movement, along with various degrees of pain. Identification of the causes of these disorders so that correction or ultimately prevention of these disorders is therefore the goal of spinal research.

However, for many cases of back pain, there are no clear physical symptoms which can be related to the underlying causes.
Degeneration of the inter vertebral disc or facets can occur at various degrees. In extreme cases of disc degeneration, tears may develop in the annulus fibres radially or circumferentially and ultimately lead to complete rupture. Disc prolapse may then occur, whereby the nucleus protrudes through the annulus and may impinge on the sensitive nerve endings in the spinal canal causing considerable pain.

Milder forms of disc degeneration are characterised by loss of proteoglycans and water content from the nucleus, and a reduction in the stiffness of the collagen fibres of the annulus. The compressibility of the disc is therefore reduced and the flexibility and load bearing capacity of the spine is decreased. Also, facet degeneration is characterised by a wearing of the articular cartilage so that the gap between the facets is widened. This results in a reduction in facet contact and increased mobility to the joint. The ability of the facets to restrain and guide the movement of the vertebrae is therefore reduced. The consequent excessive range of movement can cause ligament and muscle strains, and due to stimulation of the immense number of nocioceptors (pain receptors) can be a considerable source of pain.

While milder cases of degeneration may be a cause of back pain, abnormalities are extremely difficult to identify. In addition the complex interaction of the elements may lead to pain being experienced in a certain region which is a compensatory effect for an abnormality at another level. The association between degeneration and muscle and ligament strains with excess loading indicates the underlying causes for back pain for which no physical symptoms can be found that are necessarily mechanical (Gopal and Fast, 1997).
2.3. THE PROBLEM OF LOW BACK PAIN

The majority of back pain incidences are reported for the lumbar regions of the spine and are particularly high in Westernised societies (Kelsey, 1980). Indeed Bonica (1982) reports that primitive peoples in Africa and the Orient show a low incidence in low back pain. By contrast, in the UK in 1992, 60% of adults were estimated to suffer from back problems at some stage in their lives (National Back Pain Association). Similarly in 1991, it was estimated that 80% of the US population suffer at least one disabling episode of low back pain in their lives, and 53% suffer back pain of some sort in a year (Bonica, 1982). Also for a sample of Dutch trades and professions, of 8748 workers, 26.6% were reported to frequently experience back pain, and 1.5% (131 workers) had been absent from work due to back pain during the previous two months (Hildebrandt, 1995).

The costs of back pain are considerable. In 1982 Bonica (1982) reported in the US, data from several local regional and state surveys which implied that costs to the national economy due to low back pain were of the order of 8 billion dollars. This accounted for loss of work production, costs for health care, compensation and litigation. More recently, in 1992 costs to industry through loss of production were estimated to be at least 5.1 billion pounds in the UK, and to the National Health Service as 480 million pounds (National Back Pain Association). Prevention of the back pain would therefore be extremely beneficial to the economy of several countries.
Suffering experienced by the individual concerned cannot be quantified. The number of individuals affected are greater than the number of cases reported. Snook (1980) reported 643 out of 2000 people who had suffered from back pain had not sought help from a doctor. This is possibly because there is little treatment which doctors can prescribe when the underlying causes are unknown. In 1992 in the UK it was estimated of the 14 million GP consultations per year, 85% of cases received no differential diagnosis, and only 1% of cases result in surgery (National Back Pain Association).

Numerous studies have shown the incidence of low back pain to be related to a number of factors including gender, age, hereditary, psycho-social and work related areas. Of these factors, a strong association has been identified between low back pain and work requiring lifting activities (Snook, 1980, Jayson, 1981 and Leboueff-Yde, 1997). Hildebrandt (1995) analysed the yearly health survey from the Dutch Central Statistics Office over three successive years. Workers in non sedentary physically demanding jobs were reported to be more at risk to back pain than those in light sedentary jobs. In particular construction workers and female nurses were categories which showed high prevalence rates. Indeed Jayson reports that industrial surveys show the incidence for unskilled older manual workers is five times greater than for light work (Jayson, 1981).

There is also a high incidence for vehicle operation such as truck driving involving prolonged sitting and exposure to vibration (Jayson, 1981, and Hildebrandt, 1995), and a high incidence of low back pain in desk jobs (Jayson, 1981).
A high incidence of low back pain has also been associated with pregnancy (Bullock et al., 1987). This is thought to be linked to an increase load due to the weight of the foetus, hormonal changes causing a loosening of the ligaments reducing their ability to stabilise the joints, changes in body weight distribution and accompanying changes in curvature.

Common to pregnancy and the industrial tasks discussed are an increase in loading on the spine and constraints on spinal curvature. Manual materials handling and nursing involves the lifting of large loads in particular postures. Prolonged sitting such as in driving or in desk jobs also requires working in a posture for lengthy amounts of time and may encourage weakening of the postural muscles, poor posture and weakening of the spine. Adopting the incorrect posture during repetitive lifting tasks, or prolonged sedentary tasks may therefore be linked to the causes of back pain. A large proportion of low back pain cases may therefore be mechanically related.

To help treat patients suffering from back pain, the National Back Pain Association (1992) reports for the majority of cases, two days of bed rest was effective. This relives the loads applied on the discs and muscles. Several other studies investigating treatment for back pain have reported the initial care for low back pain should be a short period of bed rest. (Wheeler, 1995 and Twomey and Taylor, 1995). Wheeler (1995) also reports anti-inflammatory drugs may be prescribed for short term relief. However the subsequent period of recovery requires exercise to help strengthen postural muscles, and to give the individual responsibility for their own care (Wheeler, 1995 and Twomey and Taylor, 1995). There is thus no immediate cure for the problem of low back pain, and there is also no guarantee that the problem may recur.
The effectiveness of physical conditioning in helping to restore individuals suffering from back pain, strongly implies that the causes for these cases of low back pain are related to incorrect mechanics. The occurrence and recurrence of such cases could therefore be prevented if the correct postural guidelines were identified and observed. Costs to industry, medical expenses and the pain and distress to each individual could therefore be reduced. Research into the mechanics of the spine, for various postures whilst lifting various loads may therefore be beneficial to this cause.
2.4. COMPARISON OF SPINAL RESEARCH TECHNIQUES

2.4.1. REQUIREMENTS OF SPINAL RESEARCH

Everyday activities encompass a large range of postures and a number of loading conditions. To understand the normal behaviour of the spine, knowledge of the loads encountered and the loads the spine can withstand is essential. A comparison between the applied loads and the load bearing capacity of the spine may then indicate the likelihood of overuse, injury or abnormal behaviour.

During everyday activities the spine supports the weight of the upper body and any additional loads carried in a large variety of postures. Measurement of the load bearing capacity of the spine requires an estimation of the strength of each individual component responsible for resisting loads. These include the muscles, ligaments, inter vertebral disc including the nucleus and annulus, facet joints, and vertebral bone. Comparison of the estimated loads with the load bearing capacity provides an indication of the level of loading for a given posture.

Existing research into the load bearing capacity of the spine, and the loads acting on the spine for given activities involve experimental and mathematical research techniques.
Experimental methods used to determine the applied forces and the spinal configuration are necessary to provide input data for mathematical models. Estimates of body weight can be made based on cadaveric studies of the mass proportions of each major limb segment (Contini, 1963 and Dempster, 1955), or from more detailed cross sectional studies determining the weight acting at each vertebral level (Takashima et al., 1979). Also a number of methods have been used to record vertebral configuration. These include X-ray photographs (Ghista et al., 1998), video fluoroscopy (Cholewicki and McGill, 1992), skin profile measurements (Snijders and van Riel., 1987), and a lumbar motion monitor (Marras et al., 1992). The latter automatically digitises the vertebral positions into mathematical co-ordinates. However the ability for mathematical modelling and experimental techniques to predict the intrinsic forces of the spine varies considerably.
2.4.2. COMPARISON OF EXPERIMENTAL AND MATHEMATICAL MODELLING RESEARCH TECHNIQUES

Experimental procedures are based upon physical testing of cadaver specimens or living subjects, and require considerable time, expertise and costs to perform. The amount of data which can be obtained is also limited. The cadaver specimens are often of elderly people who have undergone inactivity and tissue atrophy, and the load bearing properties of the spine are reduced. Further studies on living subjects are restricted to moderate loading to prevent injury.

The most well known technique for measuring intradiscal pressure is that reported by Nachemson and Elfstrom (1970). A pressure transducer attached to the tip of a needle is inserted into the nucleus from which measurements of the intra discal pressure can be made. This can be performed on both cadaver specimens and on living subjects, although the latter is highly invasive. A significant amount of equipment and expertise is also required for this operation.

Representation of the behaviour of the spinal components by mathematical functions transforms a physical investigation into a computational problem. This allows various loading conditions to be tested without risk of damage or injury to real physical tissues. Extreme ranges of motion and the full load bearing capacity of the spine can thus be investigated using mathematical modelling techniques.

A number of further advantages associated with mathematical models make them an attractive tool for investigating the behaviour of the spine, in both a normal healthy state and when affected by disorders.
Experimental techniques are often restricted to working within the range of geometric and material properties of the available specimens or subjects. In contrast, mathematical models are constructed by ensuring the correct function and values are prescribed for each component. The range of material and geometric properties to be tested can thus be selected.

Representation of the material and geometric properties of the spinal components by mathematical parameters allows a variation in these properties to be investigated merely by changing their value. This could be a geometrical property such as a dimension, or a material property such as an elastic modulus value.

Systematic testing of each individual element by addition or elimination from the system allows a greater understanding to be gained. This can be performed using cadaver specimens, although in testing to failure for the vertebra, disc, ligament or motion segment, the same specimen cannot be used again. Consequently when experimental testing using cadaver specimens the number of variables in the system is high.
In contrast, the use of parameters in a mathematical model allows the effects of one property in particular to be observed whilst keeping the others constant. They thus allow the testing conditions to be controlled precisely. The same model may therefore be used repeatedly in tests, without the material properties changing as a consequence of loading. The biological variation between real life and cadaver specimens may be eliminated by assigning the same material and geometrical properties to the model. However representation of the spine in mathematical form requires a vast amount of computations which may be tedious and error-prone. Developments in computer techniques now enable these calculations to be performed automatically, allowing the modeller to spend more time interpreting the results. A series of mathematical investigations can be simulated using a computer overnight. Mathematical models which are created with sufficient expertise to investigate a situation do therefore not require supervision when running. Indeed Liu and Goel (1987) report computer-based models to be “the most compact and economical of all research methodologies available”.

The spine is a highly complex biological system which cannot be represented in a model without simplification. Options are to represent the whole spine with geometric simplifications, or to model part of the spine in greater detail. This depends upon the application which the model is investigating. Later developments of the model may determine the effects of additional features as they are added.
Detail of the individual components is sacrificed to be able to model the whole (or near whole) spine. Consequently modelling of the spine as an arch, a beam or a lever has formed the basis of several models investigating the overall behaviour of the spine. Alternatively, investigation of the behaviour of the individual spinal components requires that they be suitably represented. The vertebrae are often considered as non deformable under statically applied loads and are represented as rigid bodies. In contrast, deformation of the disc is considerable. Beam elements with separate stiffness properties for compression, antero-posterior shear, lateral shear, axial torsion, lateral bending and sagittal bending are often used to represent them macroscopically. The ligaments which align their fibres to the direction of the applied tensile load are thought to be suitably represented by non linear bi-axial springs.

Microscopic representation of the spinal components includes consideration of their detailed substructure using finite element techniques. Finite element models allow for the prediction of internal forces and displacements, and consequently stress and strain patterns. They can thus be used to determine the stress and strain distributions within the actual individual spinal components which neither analytical (macroscopic) models nor experimental techniques provide.
Essentially the relation between the spinal components and the applied load is determined by their material properties. Yoganandan et al. (1987) describe two types of analyses which can be performed when representing the spine mathematically.

i) A forward investigation can be performed when the applied loads, material and geometric properties and boundary conditions are known and are used to predict the resulting deformations.

ii) An inverse investigation can be performed when the applied loads, boundary conditions, geometric properties and deformations are known and are used to predict the material properties which would give rise to these deformations.

Thus using mathematical models, by ensuring the behaviour of the model agrees with that observed experimentally, data for the material properties can be generated.

Analytical models of the spine which represent the individual components of the spine by mathematical elements such as beams (discs), springs (ligaments) and rigid bodies (vertebrae) yield the material properties of these elements to the same degree of detail as that yielded experimentally. However, finite element models which allow the individual components to be represented in greater depth, provide much more detail about the substructure of these components.
The amount of data which is yielded by both types of models exceeds that of experimental investigations. Modelling techniques are not restricted by the time, expense and surgical or practical expertise required by many experimental studies. They can also be used to predict material properties of spine provided their behaviour agrees with experimental data. Also the freedom of mathematical models to investigate hypothetical situations allows them to generate a greater understanding of the response of the spine to these loading conditions. Consequently models have the potential to accelerate spinal research.

The outcome of mathematical models must be validated by experimental techniques. Unlike in vivo observations which include the neuromuscular response, mathematical modelling cannot generate the unique decisions made by the central nervous system. The dependence upon experimental measurement techniques which provide input data or validation of models which include muscles is therefore particularly strong.

Both experimental and mathematical modelling procedures have benefited from computer technology. The recording, storage and processing of vast amounts of data gathered experimentally using (for example) spreadsheet applications is particularly appropriate. However, mathematical modelling investigations can be run entirely on computers. Simulation and animation of the system using CAE applications provides immediate feedback, whilst compatibility with other applications allows rapid transfer and processing of the output data into meaningful form. Computer techniques have thus considerably accelerated the development of computer based modelling techniques. Consequently the attractive features of spinal models has resulted in them becoming an invaluable means of supplementing experimental investigations into the behaviour of the spine.
A review of mathematical models offers an insight into how models have benefited spinal research, and may indicate areas of research in which modelling techniques could be continued. It would also be an easy reference source for other researchers. However models vary widely in several features:

i) the purpose of the investigation;
ii) the particular region of the spine to be focused upon;
iii) the loading conditions to be tested for;
iv) the analysis or techniques used.

An overall review of the existing spinal models requires a comprehensive classification scheme which accommodates for this considerable variation.

In their review, Liu and Goel (1987) reported that the spinal models could be categorised as either static or dynamic, and could be further categorised as to whether they could accommodate large displacements, whether the vertebral bodies were treated as rigid or elastic and according to the techniques used to models the disc and ligamentous structures. They further discuss these models under whole spine, head-neck models, lower-back models and disc-body unit models categories. However, this review covered the development of models of the 1980's, and a considerable number of models have been developed since.
Yoganandan et al. (1987) categorised the models into four classes:

1) Nature of external loading (static, dynamic, kinematic).
2) Geometric (single structure or individual components).
3) Type of analysis:
   a) to input geometry, material properties and boundary conditions and to predict deformations;
   b) to input geometry, boundary conditions and deformations and predict material properties.
4) Application (mechanical or clinical).

However, the authors reported that these classification groups overlap each other.

More recently, Goel and Gilbertson (1995) looked at the application of finite element models to clinical problems. Their categorisation involved simple, detailed or a combined approach to the finite element models. The simple models represented a large portion of the spine such as thoraco-lumbar spine, but each motion segment with only one or three elements. In contrast the detailed models represented only one or two motion segments, but with much greater detail for the individual components of the spine, such as disc and ligaments. This latter category included modelling of the trabecular bone in the vertebral body, vertebral bone remodelling, the disc annulus composite, and the poroelastic behaviour of the disc.
Each spinal region demonstrates variation in material and structural properties which reflects their distinguished roles in each region. Due to these structural and mechanical differences, each region is linked to characteristic disorders. A classification scheme which allows models of each region of the spine to be compared, therefore allows models of similar application, loading conditions, and region to be compared. The contribution of the models under a common application can therefore be evaluated.

A comprehensive review of the existing mathematical models of the spine has been completed (Grilli and Acar, 1997c) and elaborates on the classification scheme used by Liu and Goel (1987) in considering whole, partial and unit spinal models. The classification is as follows:

1) Whole spinal models
These models focus upon the gross spinal behaviour, with simplification in geometric representation.

2) Partial spine models
To investigate the behaviour of a certain section of the spine in greater detail. These may be investigating characteristics of a certain region:
   i) head and neck regions;
   ii) thoraco-lumbar regions;
   iii) lumbar regions.

3) Unit models.
To investigate the behaviour of motion segments, and smaller units such as single vertebrae in microscopic detail. These again possess material and geometric properties from each of the three regions of the spine:
   i) head and neck regions;
   ii) thoraco-lumbar regions;
   iii) lumbar regions.
The review attempts to illustrate the benefits of spinal models to society on a macro and microscopic level using this classification scheme. There are five main ways in which spinal models have contributed:

1) Models can generate a greater understanding of the behaviour of the spine and its components.
2) Models can be used to determine the main components of the spine which influence spinal behaviour.
3) To investigate the effects of certain disorders on the behaviour of the spine.
4) To investigate the causal mechanisms of these disorders.
5) To evaluate and supplement clinical correction techniques.

The review is presented here to illustrate how models have benefited spinal research and to indicate areas of research in which modelling techniques could be continued. Particular focus is on models developed to investigate low back pain due to the prevalence of this problem.
2.5.1. WHOLE SPINE MODELS

Early mathematical models of the spine were concerned with investigating the overall behaviour of the spine. Whole spinal models were therefore developed. These models are inherently limited by the complexity of the overall structure of the spine, although computational power and limited data available from experimental procedures for input or validation have also been restricting.

One approach was the development of continuum models which reduced the geometric complexity of the model. This allowed more computational power to be used to investigate dynamic situations, such as the 'pilot ejection problem' and vehicle collision which were associated with a high incidence of spinal injuries. Several of these models have previously been reviewed by Yoganandan et al. (1987)

As an example, Cramer et al. (1976) produced a one dimensional continuous rod model which represented the whole spine as a homogeneous beam column with infinite degrees of freedom. This included the initial sagittal curvature of the spine. The model was used to investigate a pilot ejection situation, by simulating an acceleration force (10 g) applied at the pelvis in the caudo-cranial direction. In addition, the weight of the trunk was applied as a distributed force acting over the entire length of the spine anterior to the vertebral centres. This data was obtained from cadaver specimens. The axial force, shear force, and sagittal bending moment at each vertebral level were then considered. The relations between these forces and the resulting stress were assumed linear, and the resulting strains small. The model was however non-linear, accounting for large deformation of the spine. The governing equations were therefore solved using a finite difference method and iteration techniques.
The outcome of the model showed the configuration of the spine, the axial and shear force and bending moments for the impact situation at various time intervals. An example of the configuration history is shown in Figure 2-2. The authors report the model predictions to agree well with pilot ejection injury data, although viscoelastic properties of the spine, and the modelling of external constraints such as seat back and shoulder harness was omitted.

Figure 2-2: The configuration history of the spine predicted by the model. (u is the horizontal displacement from the vertical reference axis, w is the vertical displacement of the spine, \(w_0\) is the initial vertical displacement of the spine and z is the co-ordinate along the vertical reference axis.)

(Taken from Cramer et al., 1976.)
Two dimensional models were also developed for the whole spine. A continuous two dimensional model developed by Soetching (1973) was the first to include the stretch reflex response of the muscles to investigate the flexural motion during vehicular collision in the sagittal plane. Lindbeck (1987) also developed a continuous two dimensional model to predict the flexural rigidity of the spinal column in the frontal plane, a procedure extremely difficult to perform in vivo.

Unlike continuum models, discrete parameter models represent the spine as a series of individual components such as vertebrae, inter vertebral discs, and ligaments. This has encouraged more detailed modelling of spinal components. The two dimensional (2D) discrete parameter model developed by Orne and Liu (1971) was of particular significance as it was the first to allow for shear and bending moments as well as compression. It allowed for variation in size and mass of the elements at each level, and included viscoelastic properties of the disc, the effects of rotational inertia of the vertebrae. Also the analytical formulation was non linear, allowing relatively large deformations to be described. The large motion of the spine during impact could thus be accounted for. The model was successful in contributing to the explanation of the high occurrence of anterior lip fractures in the vertebrae associated with the bending of the spine in pilots ejecting from high performance aircraft. However, the model was limited by the lack of data for the shear and bending material properties.
Belytschko et al. (1978) later produced a detailed three dimensional model of the whole spine in order to investigate the pilot ejection problem. The vertebrae, head, pelvis and ribcage included in the model were each represented as rigid bodies. The thoracic and lumbar vertebrae were each interconnected by a beam element representing the inter vertebral disc, and by seven pairs of spring elements representing the spinal ligaments. The variation in axial, torsional and bending stiffness of the disc was defined (taken from the literature) accounting for variation from level to level. The cervical spine was represented in a similar manner, but only the interspinous ligaments represented by one pair of spring elements were included. The articular facets were represented by hydrodynamic elements which were governed by fluid-pressure laws. These were considered to represent the directional properties of the facets more accurately than spring elements.

Each rib was connected to two vertebrae by a costo-vertebral joint, which was modelled by deformable elements. The data for this rib-cage model was taken from the literature. A simplified model was also developed, representing the ribcage and viscera as a column of deformable elements parallel to the spine. This increased the computational efficiency by a factor of 10, so that parameter studies could be performed.

Equations of motion for the system were derived, and the analytical procedure accounted for non-linear behaviour. This included large three dimensional displacements and rotations of the rigid bodies, and non-linear stiffening behaviour of the deformable elements. The model is shown in Figure 2-3.
The ejection system modelled included the pilots' chair represented by a vertical and a horizontal plane surface, and a restraint system represented by three springs connecting vertebrae T1, T2 and T3\(^1\) with the seatback, and a single spring connected to the pelvis. The pilots' helmet and helmet mounted devices were assumed to move with the head.

The model investigations included parameter studies (using the simple model) to investigate vertical ejection with slow and fast onset of acceleration (rate of ejection) and the angle of ejection to test the effects of varying the seat angle. The effects of a helmet mounted device were investigated by increasing the mass and eccentricity of the head in both sagittal and frontal planes. The results of the model used to simulate the case for vertical ejection with slow rate of acceleration were also reported so that a validation between the simple and complex models could be obtained. The model predictions provided a history of axial force and flexion moment for various levels of the spine and the head which could be compared against known values of failure for the spine.

\(^1\) T1, T2 and T3 refer to the first three thoracic vertebrae as illustrated in Figure 2-1.
Models of the whole spine have thus mainly been used to investigate dynamic behaviour of the spine during impact situations.

A two dimensional static model developed by Sierig and Arivikar (1975) included muscle forces represented by force vectors acting over the whole spine. The individual vertebrae of the lumbar and cervical regions were each represented by rigid bodies, while the thoracic region was represented collectively as one single rigid body, reflecting the limited range of motion these vertebrae undergo. Equations of equilibrium for each inter vertebral level were defined and were solved using an optimisation procedure. This returned a solution which minimised the sum of all muscle forces, joint reaction moments and joint tensile reactions. Thus for a given posture the model could be used to predict the lowest joint reaction and muscle forces which satisfied equilibrium at each level. The authors reported excellent agreement between predicted inter vertebral disc reaction forces and experimentally measured values at the L3 and L4 level. However the muscle force recruitment strategy was not validated with electromyographic (EMG) data.

Three dimensional static models have been developed, but simplified due to the computational requirements. The three dimensional (3D) model by Schultz and Galante (1970) was restricted to a purely geometric analysis. The model was thus useful in demonstrating the motion of the spine, and included flexion, extension, lateral bending and twisting and was later applied to the clinical case of scoliosis. The later model of Panjabi (1973) included a three dimensional force-deformation relation for adjacent rigid bodies interconnected by deformable elements which could be used to describe the displacement of the spine under loading. This force method has been used in many successive models.
The overall motion of the spine is determined by the cumulative motion of the individual vertebrae, with coupled motion between them. An analysis of the motion at the individual spinal element level thus provides a more detailed understanding as to how the spinal mechanism functions. Models of smaller regions of the spine have allowed for this.
2.5.2. PARTIAL SPINE MODELS: HEAD AND NECK MODELS.

In the cervical region, the vertebra must support the immediate weight of the head, and the large moments it generates when acting eccentrically. In impact situations, the inertia of the head results in considerable eccentricity and a number of cervical injuries have been reported. Several models of this region have therefore been concerned with investigating the mechanisms of these injuries, so that possible prevention strategies could be proposed. Many of these involve representing the vertebrae as rigid bodies, and the discs as a combination of springs and dampers, each with axial, torsional and shear responses to account for viscoelastic properties. However, the inter vertebral discs within this region are relatively thin and to assist in the load bearing, the facets play a large role under moment bending. Prasad and King (1974) were the first to demonstrate the interaction of the facets and disc as load bearing paths through a simulation of an impact situation for a vehicle seat occupant using a 2D model of the head and neck.

Williams and Belytschko (1983) later reported that previous models which use springs to represent the facets underestimate their load bearing role, since springs do not have shear stiffness and result in large unrealistic shearing displacements between vertebrae. This is particularly important for the cervical region in which the facets play a significant role in load bearing. Thus they developed a more elaborate representation of the facets, with axial and shear stiffness allowing simulation of frontal and lateral plane and therefore three dimensional motion. This was consequently the first three dimensional model of the head-neck to be developed, and was used to investigate the effects of impact acceleration.
The vertebrae C1 to T1 and the head were modelled by rigid bodies, interconnected by deformable elements. These included spring elements with axial stiffness only representing the ligaments; beam elements with axial (different in tension and compression); bending and torsional stiffness representing the inter vertebral discs; pentahedral elements representing the facet joints; and muscle elements.

In total, twenty two muscle groups were included. The force each muscle group could produce was estimated from the product of its stress and cross sectional area. The stress-strain relation for the element included a viscoelastic parameter as well as elastic moduli, all of which were dependent upon the concentration of a molecule within the muscle cell defining its level of activation. To represent the stretch-reflex response of a muscle during impact, upon stimulation of the muscle cell a constant influx of the molecule to the cell occurred. Before this, the concentration of the molecule was constant (zero influx). The model was thus the first to include both passive and active contractile properties of muscles.
Simulations for impact with and without muscle activity were performed so as to observe the effects of the muscle stretch-reflex response during impact. The model predicted muscle contraction forces to decrease the force in the ligaments upon both frontal and lateral impact. Also, with frontal impact compression, shear forces and bending moments in the disc were decreased with muscle activity, while tensile forces in the disc and forces normal to the plane of the facets increased. In contrast, under lateral loading the stretch reflex response increased peak compressive and shear forces in the disc while peak tensile forces stayed approximately the same. This was in closer agreement with the experimental results for living subjects than the model without muscle forces. The study therefore emphasised the importance of the muscle stretch reflex response in impact simulations and showed the different responses under lateral and frontal impact, emphasising the importance of considering three dimensional motion under impact.

The three dimensional model by Merrill et al. (1984) also included muscle forces in order to investigate the response of the head and neck to both frontal and lateral impact. However, in this study it was assumed that activation of the muscles would be too slow to affect the response time in this impact study. The passive stiffness properties for each muscle and ligament (relating force to elongation) were therefore defined. The study investigated the response of the model to short term impact. This included a 2D flexion whiplash due to base acceleration, frontal skull impact and a blow to the side of the skull. Numerical integration methods were used to determine linear and angular displacement, velocity and acceleration, facet and joint loads, and muscle force history. However, the authors did not find good agreement between the model predictions and experimental studies performed on living human volunteers.
Deng and Goldsmith (1987) developed this model to include a better representation of the inter vertebral joint and muscle forces. While the model of Merril et al. (1984) included 7 pairs of muscle forces assumed to act in a straight line between their points of attachment, Deng and Goldsmith (1987) included 15 muscle pairs acting along a path defined by three nodes so that the line of action could account for curvature in passing around bones and soft tissue. The model is shown in Figure 2-4.

Figure 2-4: Numerical model of head/neck/upper-torso, showing the attachment points of the muscles. (Taken from Deng and Goldsmith, 1987.)
The inter vertebral joint was modelled in three dimensions. However, stiffness properties of the joint had only previously been determined for single motion segments. The current study therefore estimated joint stiffness at all levels based upon the proportion of the cross sectional areas of the discs. Also the authors report the numerical method 'Adams-Bashforth-Moulton' scheme was used to predict displacements from the assigned force and stiffness values, resulting in higher accuracy than the previous model. The model investigated the effects under frontal impulse loading (flexion whiplash), lateral impulsive loading, and simulated cases tested by a previous physical model including impact to the head.

The authors report the more efficient numerical scheme employed in this model resulted in less computing time and involved higher accuracy. In addition, the discs were represented in a fully three dimensional manner, and the muscle lines of action were presented more precisely by three point elements rather than two. As a result, the model compared favourably with the experimental cases simulated. However, for the flexion whiplash case, the model predicted the head to return to its neutral position whereas the experimental case (with living subjects) showed little rebound of the head. The authors report this might be different were the active component of muscle forces considered rather than just passive behaviour.
More recently, Hoek et al. (1993) included the neck muscle forces to investigate the influence of helmet mounted devices on the cervical spine of F-16 pilots. A video camera was used to record the position of markers attached to the head and shoulders of a pilot during four flight operations. Three accelerometers attached to the helmet measured the three components of direction and magnitude of the acceleration load during the flights. It was assumed that in F-16 flights, the forces in the neck due to head movements were negligible in comparison with the load contributions of gravity and acceleration of the aircraft. A static analysis was therefore performed involving this acceleration load vector, and inertial properties of the head were ignored.

Digitisation of the markers allowed the position of the pilot to be reconstructed, based on the three dimensional model of Snijders et al. (1991). The origin of the model was at T1, so that motion of the head and neck was described relative to the trunk. Flexion was possible at the atlanto-occipital joint, C1-C2 joint, C2-C3 joint and C7-T1 joint. However, for simplicity, the cervical vertebrae C3-C7 and their discs and muscles were incorporated as a single rigid link whose length adjusted as a function of its flexion angle. This accounted for bending and stretching of the cervical spine, although individual joint forces in this region could thus not be calculated. The model is shown in Figure 2-5.
Figure 2-5: Side-view of the neck model. The ventral side is at the right. TC is the centre of gravity of the head, PO is the point of attachment of the dorsal neck muscles. The points A-E and O are centres of rotation of the links. The positive directions of the X-, Y- and Z-axis and of the rotations α, β, and γ are according to the right-handed co-ordinate system shown. (Taken from Hoek et al., 1993.)

The number of muscle forces involved produced a statically indeterminate system. Only three muscle forces were required to equilibrate the three components of joint reaction force in the atlanto-occipital joint, and ensure the head was kept in a static position. Therefore the muscle forces and joint reaction forces were calculated for every combination of three muscles, and the solution which led to the smallest joint reaction force was chosen (provided muscle forces were positive, i.e. not in compression).
A total of 330 postures were analysed from the flights performed. The loads calculated on the cervical spine were compared with known failure data for the loads on the neck vertebrae, while muscle forces were compared with known maximum muscle strengths. Simulations of various helmet mounted devices were also performed by summing the gravitational forces acting on the mass of the helmet and helmet mounted devices and the head into one load vector acting through their common centre of mass. The effects of a posterior counter weight (to offset the anterior eccentricity of the helmet mounted devices) was also investigated, along with a simulation for no helmet or devices. Their influence on muscle force requirements and joint reactions could thus be observed. In this way the model provided a detailed analysis of impact situations, the results of which could help to improve seat and helmet designs, and to reduce the incidence of fractures. However, reactions at the joints included in the single link (C3-C7) could not be determined.

Finite element models of the cervical vertebrae have been also been developed and have been reviewed by Yoganandan et al. (1996). These models exploit the finite element methods developed from modelling of the lumbar vertebrae.
Kleinberger et al. (1993) developed a three dimensional finite element model of the ligamentous cervical spine to investigate the mechanics of automotive crashes and identify possible causes of injury. However, this represented the geometry of the vertebrae by regularly shaped figures. Yoganandan et al. (1997) later reduced the number of vertebrae modelled to three segments to allow for a more detailed geometrical representation based on CT scans. This included C4, C5 and C6 vertebrae (Figure 2-6). The model was tested under uniform axial compressive loading, and included a parametric study to determine the effects of varying the material properties of the vertebral components under axial compression. This included the material properties of the nucleus, annulus, vertebral body core and shell, posterior elements and end plates.

Figure 2-6: Finite element model of cervical spine structure.
(Taken from Yoganandan et al., 1997.)
Some simplifications in this model included representation of all elements with linear isotropic and homogenous properties, representation of the annulus as a single structure rather than as a composite fibre-matrix and a simple representation of the facets, rather than using contact elements employed in previous lumbar models. In addition, lack of material property data for the cervical region required data from other regions to be assigned. Despite this, the model showed good agreement with experimental data. The authors report the particular feature of this model was the inclusion of three motion segments which provide realistic physiological boundary conditions for the middle vertebra; whereas previous single or two level models require constraints which result in non physiologic conditions.
2.5.3. PARTIAL SPINE MODELS: THORACO-LUMBAR REGION

The cervical spine is mainly responsible for the motion and support of the head whereas the thoraco-lumbar spine is responsible for movement and support of the trunk and for bearing the weight of any external loads carried. More detailed investigations into the movement and weight bearing functions of the spine have therefore focused upon the thoraco-lumbar regions. Models developed in this area have involved continuous and discrete parameter representations and also finite element techniques.

Belytschko et al. (1973) developed a three dimensional discrete parameter model of the thoraco-lumbar spine. The vertebrae were represented as a series of rigid bodies, interconnected by deformable elements representing the inter vertebral discs and ligaments. The spring elements used to represent the ligaments were assigned axial stiffness properties, while the beam elements used to represent the inter vertebral discs were each assigned axial, torsional, shear and bending stiffness properties. The values for these stiffness properties were chosen from the literature reported for cadaver specimens and were adjusted from level to level according to geometric properties. The stiffness properties for the disc in flexion-extension and lateral bending were assumed equal due to the lack of experimental data available. Despite this, the authors felt their representation of the disc using four stiffness elements was an improvement on those which had previously used just one element. The force-displacement relationships for each deformable element were then defined based upon these stiffness values.
At each stage of loading, an equilibrium configuration was sought for which the external loads applied to the system were equal to the internal forces of the system. Deformation of the spring and beam elements was assumed to be small so that a linear force-displacement relation could hold. However the equilibrium equations were solved using an iterative procedure which accounted for large overall motion of the spine. Consequently the authors reported this model to be the first of its kind to include a non linear force analysis.

A refinement of this model by Schultz et al. (1973) included a more precise description of the ligaments, with the longitudinal ligaments, intertransverse ligaments, interspinous ligaments, and the ligamenta flava each represented by separate pairs of spring elements. In addition, a spring element connected the right and left articular processes of adjacent vertebrae and represented the articular facets; those in the lumbar regions were oriented posteriorly whereas those in the thoracic regions were oriented laterally. In this way, variations in the direction of the facets and their kinematic restraints could be accounted for.
Takashima et al. (1979) developed these models to investigate the effects of muscle forces on the behaviour of the trunk. Each muscle included in the model was assumed to act along a line between nodes running from its point of origin to insertion on the vertebrae. The force vector for each muscle was described in local and global co-ordinates so that a change in position of the vertebrae to which it was attached also accounted for a change in the direction of the vector. These muscle forces were based on estimates of their cross sectional areas taken from anatomical drawings (Eccleshymer and Shoemaker, 1911). In total, 68 muscle forces were included. To match the detailed pattern of muscle forces, body segment weights were estimated for each level of the trunk. For equilibrium, the forces due to the external loads were required to equal the forces due to the deformation of the soft tissues and the active contraction of the muscle forces.

Also, unlike the previous models which were limited in data for the stiffness properties of the disc, the discs in this model were represented by six stiffness parameters, describing the disc resistance to axial deformation, torsional rotation, lateral bending, antero-posterior bending, lateral shear and antero-posterior shear.

The model was used to determine the effects of body weight on the maintenance of three upright postures, and to determine the effects of bodyweight and muscle forces on a scoliotic curve. The muscles were assumed to control spinal posture so that spinal displacements were small. This justified the assumption of linear geometric and material properties. The model predictions for disc compression and restored scoliotic Cobb angle with muscle contraction forces agreed fairly well with other studies.
The value of these models is reflected in their later applications into the study of the clinical disorder of scoliosis and the effectiveness of correction techniques. In particular, Schultz et al. (1981) performed a study to predict the effectiveness of electrical stimulation of the muscles for correction of scoliotic spines.

Inclusion of the rib cage by Andriacchi et al. (1974) required modelling of the sternum and bony parts of each rib using rigid bodies and the intercostal spaces and costal cartilage using deformable elements. Additional points were added to each vertebra to define the location of the costo-transverse and costo-vertebral articulations. These were located according to descriptions in the literature. In addition, rib articulation points were defined on the rigid body representing the sternum so that 7 rib pairs were attached. The geometry for the sternum, and for the ribs was again obtained from the literature. The model is shown in Figure 2-7.

Figure 2-7: Computer generated views of the lumbar spine and skeletal portions of the thorax in a normal anatomical position.
(Taken from Andriacchi et al., 1974.)
Determination of the material properties of the ribs was also required. Experimental tests on the deformable properties of isolated cadaveric ribs had previously been reported in the literature. A test for rib 6 was simulated on the computer, and the joint stiffness values adjusted until the model predictions for displacements closely matched the experimental values reported. Repetition of the test for other levels showed the difference in displacement magnitude at each level was related to the difference in articulation geometry. The same stiffness values for the joints at each level was therefore assigned.

The model was then used to investigate the effects of the rib cage on the bending response of the spine. Both an isolated spine and one with the ribs attached were therefore simulated. The model predicted that the rib cage increased the overall stiffness of the thoraco-lumbar spine and also the stiffness of the motion segments at each level of the thoracic spine. This increase in stiffness implied greater stability of the spine.

Wynarsky and Schultz (1991) developed the existing computer model of the spine with the rib cage and muscle forces in order to investigate the correction of scoliosis by bracing techniques, but included revised locations of muscle origins, insertions and lines of action. The brace forces, simulating the behaviour of a Boston brace, were applied as 196 potential point loads evenly distributed over the patient-orthosis surface. The inclusion of muscle and brace forces whose values were unknown resulted in a highly indeterminate set of equations for the system. A non linear optimisation procedure was therefore used.
The study determined the patterns of bracing forces which gave the optimal correction of the scoliotic configuration. Both passive bracing forces and muscle forces were found to have corrective effects even at non maximal level on Cobb angle and lateral alignment, although small left lumbar and left high thoracic secondary curves were created. The appearance of these secondary curves had also been reported in clinical observations, thereby providing some validity to the model. Model predictions also implied that scoliosis is not likely to be fully corrected by passive bracing or muscle contraction, although the muscle forces were predicted to be more effective than the bracing forces.

A two dimensional model of the thoraco-lumbar spine developed by Ghista et al. (1988) was used to investigate the surgical correction of scoliosis using a Harrington distraction rod, and to predict the optimum combination of distraction and lateral traction forces needed to be applied.

The geometric properties were simplified, with the overall motion segment represented as a cylindrical beam which could resist axial, shear and bending moments. Variation in material and geometric properties from level to level was included. However, it was therefore assumed the motion segment was a homogeneous, isotropic structure. The focus of the investigation was thus not on the individual effects on the discs and vertebrae, but rather on the overall curvature of the spine.

The scoliotic curve was developed in the model by applying nodal forces until the curvature matched those of radiographs of clinically diagnosed cases. The stiffness at each level of the spine would then be determined using finite element procedures, with the relation between nodal forces and displacements of each beam defined in matrix format.
The model was then used to simulate the surgical corrective forces required to rectify the deformation. This was determined by applying a set of forces in stages to the model and observing the deformation response, until the desired restorative deformations were obtained. The increase in stiffness response of the spine to increased traction required a non linear analysis. However, the finite element model was linear elastic allowing only small displacements. An incremental approach was therefore used, which allowed the behaviour at each stage to be linear, but which produced an overall non linear response.

An intraoperative modelling system including strain gauges was then developed to monitor the effects of the forces predicted by the model as they are applied during surgery.

Despite the simplification in geometry and material properties, good correlation was achieved between surgical and model spinal curves. The model is thought to be the first to be applied to assist with surgical correction in this way, and indicates the potential that models have in assisting with spinal disorders.

An investigation of the curve progression in idiopathic scoliosis was performed by Patwardhan et al. (1986) using a simple continuum model. In particular, the effect of curve pattern and curve magnitude on spinal stability was considered. The spine was represented as a flexible beam, initially curved in the frontal plane. This curvature was approximated by a linear combination of trigonometric functions. Selection of the number of terms and the values for coefficients in this series allowed the initial geometry of any scoliotic curve pattern to be simulated. For simplification, only two terms were used. This was justified by the close agreement found between the computed Cobb angles for these mathematical curves and those of radiographic measurements of scoliotic spines.
The critical load for the initially curved beam was defined as the load at which progressive plastic deformation would occur. Model predictions for the critical load were used to express the load bearing capacity of a scoliotic spine relative to that of a normal one. This was performed by determining the ratio of the critical load for the curved beam to the Euler buckling load for an initially straight (normal) beam. Results plotted for a number of curvatures showed how the load bearing capacity of the spine was affected by a scoliotic curve.

The effects of an orthotic correction system for scoliosis were also simulated in the frontal plane. Transverse loading was applied, and boundary conditions which restricted the motion of the spine at the ends were prescribed. This is illustrated in Figure 2-8. Parameter studies were performed to determine the effect of the distribution and magnitude of the transverse load and the boundary conditions at each end of the spine. Thus the model had the potential to investigate the effects of any orthosis, including Milwaukee and Boston braces.
Figure 2-8: A biomechanical analog of a general spine-orthosis system. $L$ is the length of the thoraco-lumbar spine, $P$ is the applied axial load, $Q_o(x)$ the transverse load applied by the orthosis, and $k_i$ and $k_s$ are spring constants used to define the boundary conditions for the system.

(Taken from Patwardhan et al., 1986.)
A simple continuum model of the thoraco-lumbar spine was also developed by Noone et al. (1991) in order to investigate the progression of scoliosis for a patient with Duchenne muscular dystrophy. Characteristics of this disorder involve asymmetric muscle contractions and a weakening of muscle force over time. Asymmetric right and left muscle force pairs were therefore applied. Also, the bending of the spine due to force imbalance and a reduction in muscle force were updated at each stage of analysis.

While many authors have considered scoliosis to be a disorder in the frontal plane, Stokes and Gardner Morse (1991) suggested that it could be caused by three dimensional coupled motion of the vertebrae. A three dimensional finite element model of a ligamentous thoraco-lumbar spine was thus produced. The geometry was idealised by giving the motion segments equal lengths, represented by a nodal point on the centre of the vertebral body, while the inter segmental links were each represented by stiffness matrices. The possible causes investigated included:

1) The coupling movements within a motion segment
These were investigated by applying the moments and forces that would produce the characteristic scoliotic curvature.

2) Posterior tethering by the soft tissue during spine growth
This was simulated by adding spring elements to connect the vertebral processes on one side of the model. The tether could be shortened by applying initial strains to the springs. However, the results suggested neither hypothesis was a suitable explanation. The model is shown Figure 2-9.
Figure 2-9: Back, lateral and plan views of the model of an idealised thoracic and lumbar spine. There are 17 equally spaced vertebral nodes (shown as squares) in the form of a sine wave, with vertebral processes attached to each node. In the tether model, spring elements (shown as dashed lines) linked adjacent postero-lateral processes on the right side. These springs were omitted from the coupling model.

(Taken from Stokes and Gardner Morse, 1991.)

Mathematical models have been considered extensively in the investigation of scoliosis. Other applications of thoraco-lumbar spine models include investigation of the effects of manipulative therapy (Lee et al., 1995) and a kinematic model (Monheit and Badler, 1991) to provide a more realistic representation of human motion in the sagittal plane.
Yettram and Jackman (1982) developed a model in order to investigate the muscle and joint reaction forces associated with spinal movement in the sagittal plane. In particular, it was used to assess the changes in forces acting on the spine, when the body moves from one posture to another, brought about by muscle contraction.

A structural analysis was first performed, determining the deformation response of the soft tissue elements of the spine. The vertebrae were considered rigid, whilst beam elements were used to represent the bending characteristics of the disc and bar elements to represent the axial stiffness of the ligaments. Muscle forces were assumed to act at their points of attachment. An optimisation procedure was used to obtain the muscle and inter vertebral joint reaction forces for an initial configuration of the spine. The change in configuration of the vertebrae with posture was obtained from displacement data from radiographs. The inter vertebral joint reactions for the new configuration were then calculated in a structural analysis using elastic deformation data and the muscle forces for this position were again determined using an optimisation procedure.

The model thus accounted for body weight, external loading, muscle forces, inter vertebral reactions and disc and ligament elasticity. However, in the optimisation procedure for the final position, an infeasible solution was obtained and for the spine represented in this way, no equilibrium configuration could be found.
Other models have been used to investigate the effects of sagittal curvature on stability of the thoraco-lumbar spine (Meakin et al., 1996 and Scholten et al., 1988). The previous finite element model by Shirazi Adl and Parnianpour (1993) for investigating the stability of the lumbar spine (discussed in lumbar region models) was later developed (Shirazi Adl and Parnianpour, 1996) to consider stability of the ligamentous thoraco-lumbar spine. The effects of flexion moments generated by the action of body weight acting anterior to the spine was investigated. In addition, the effect of pelvic rotation was considered. The model was three dimensional, predicting displacements in both lateral and sagittal planes under the prescribed loading conditions.

Loading was incremental, and included an axial compressive force uniformly distributed on all vertebrae T1-L5 with that at T1 four times greater to account for the weight of the head. A concentrated load applied at T1 was also considered for comparison. In addition, several stabilising loading conditions were investigated:

i) Flexion moments were prescribed at L1 (along with the axial compression all along the spine).

ii) L1 was restrained in the sagittal direction and the sagittal moments required to stabilise the spine under axial compression were noted.

iii) Rotations at all levels were restrained in both planes and the moments required to stabilise the spine under axial compression were noted.

iv) Axial compression was applied, and sagittal rotations at S1 were varied to represent pelvic rotation. The optimal pelvic rotation which increased the load bearing capacity of the spine under axial compression was sought.
The study showed (as for the lumbar spine) that the ligamentous thoraco-lumbar spine did not buckle at certain loads, but rather showed a significant decrease in stiffness. The presence of flexion moments showed significant increase in stiffness, thereby increasing the stability of the spine.

Under axial load, the primary displacements were predicted to be in the sagittal plane. An increase in coronal displacement was found to reduce the amount of compression the spine can support. Thus, for the clinical case of scoliosis, the load bearing capacity is less according to the current model predictions. However, the presence of sagittal thoracic and lumbar curves showed less displacement in the coronal plane, thereby stiffening the spine and allowing it to support greater axial compression with minimal displacements.
The majority of models of the lumbar region have been developed to investigate the problems of low back pain. Many of these represent the spine as a lever, whereby the spine is assumed to bend about a fixed base, namely the sacrum. The flexion moments produced by body weight and loads lifted acting anterior to the spine are calculated. The model then determines the extension moment required to be produced predominantly by posterior muscles, so that flexion and extension moments are in equilibrium. These moments are generally considered about the lower lumbar levels, (L5/S1 or L4/L5) at which moments and loads are predicted to be greatest. Equilibrium of the structure is ensured if both forces and moments are in equilibrium. For equilibrium of forces, a resulting reaction at the inter vertebral joint, with compression (C) and shear (S) components occurs. Figure 2-10 illustrates this concept for a lifting activity, with the force due to body weight represented by a single force (W), load lifted (L), and a collective muscle force (M).

Figure 2-10: Illustrating the concept of the 'lever model'.
The geometry of the lumbar vertebrae and discs, and in particular their large cross-sectional area, reflects their role in bearing the cumulative loads of the upper body. The loads are therefore greatest in the lumbar regions and the effects of these weights can be further magnified when the load is held far from the body, generating large moments at the lumbar level. This is typical of tasks involving heavy lifting, such as manual materials handling, or lifting and turning patients in nursing, and may expose the spine to sudden loads of large magnitude, near to, or exceeding the individuals capacity. The model is therefore suitable for investigating the loads on the lumbo-sacral regions of the spine. Thus while effectively whole body concepts, these models have been classified in the lumbar region.

This model has been used to investigate the moments generated on the spine whilst seated at a workplace. Andersson et al. (1980) assumed in this scenario external moments correspond to the resultant stress on the spine. The arm positions which corresponded to the least moments acting on the spine could then be predicted. In this way, mathematical models could be used to recommend the correct seat height and position at a workplace. However, the overall focus has been on lifting activities performed from a standing position.

2.5.4.1. Single Muscle Force Models

The indeterminate nature of the spinal system, in which the number of muscle forces to be determined exceeds the number of equilibrium equations to be solved, has resulted in several early models representing the muscles as one collective muscle force. Sagittal symmetry is therefore assumed. This provides a simple means for investigating the loading on the spine for industrial lifting tasks.
Garg and Herrin (1979) developed a model to investigate the effects of a stooped or squat lifting technique (Figure 2-11) on the stress predicted at the L5/S1 inter vertebral joint. The authors predicted for the stooped posture which involved a trunk flexion angle of nearly 100 degrees, the flexion moments were greatest. Greater muscle force would therefore be required for equilibrium of this posture. However, the model also predicted the component of compression due to loading was lower than the component of shear. The shear forces acting on the spine during lifting are therefore a significant loading factor.

![Diagram showing stooped versus squat posture for load placed close to the body.](Taken from Garg and Herrin, 1979.)

However, this simple anatomic representation of the muscle forces has since been shown to predict unrealistic values for inter vertebral joint compression. For example Chaffin (1969) predicted a compression value of 5664N on the L5/S1 disc which lies within the reported failure values for vertebral endplates (Farfan, 1973), whereas subjects performed the lift successfully in vivo without injury.
Whereas static models predict the forces required for a person to hold a load in a given posture, dynamic models include the inertial and acceleration forces associated with lifting movement.

A six segment model of the whole body developed by Wood and Hayes (1970) to estimate the loads acting on the L4/L5 inter vertebral joint during a lifting activity predicted the inertial forces during the critical phase of motion to assist the muscle forces by resisting flexion, although these values were low due to the relatively slow angular motion. In contrast, the gravitational forces assisting flexion were much higher, requiring a greater muscle force for equilibrium, and a consequent increase in compressive force for the dynamic loading. In comparison with a static situation also simulated, the predicted compression at the inter vertebral joint was much higher for dynamic loading.

Lifting techniques have again been investigated using dynamic models. In a study by Frievalds et al. (1984) subjects were free to select the greatest weight they could lift, placed within a box of various sizes, with or without handles. Stroboscopic photographic techniques were used to record the movement of joint markers attached to the subject during the lift. Data for joint angles was collected by digitising the stroboscope recordings at each time interval. The displacement and velocity for each segment was used as input to a two dimensional model, from which ground reaction forces, reactive torques at each joint, acceleration of each link, the muscle force required, and the resulting compression and shear components of the L5/S1 joint reaction force were determined. The model predicted the compressive force pattern during the lift also showed an oscillatory effect, being high at the start and ends of the lift. This was due to the large moment arm of the load as the subject reached out to pull the load into the chest at the start of the lift, and as the subject placed the load on the bench and pushed it away at the end of the lift.
Leskinen et al. (1983) studied subjects lifting a 15kg box with handles from a shelf 10cm above the floor, using various techniques:

i) a back lift: straight knees, hips flexed and back bent
ii) a leg lift: flexed knees, hips flexed and back straight
iii) load kinetic lift
iv) trunk kinetic lift

For the kinetic lifts, the load was moved horizontally or the trunk vertically to gain kinetic energy. These techniques are illustrated in Figure 2-12.

![Figure 2-12: The lifting techniques of the study. (Taken from Leskinen et al., 1983.)](image)

Due to the dynamic effects, and the exclusion of force due to intra abdominal pressure (IAP), the peak L5/S1 compression force predictions for the leg lift and back lift were 70% and 100% higher respectively than predictions from a similar static analyses of Garg and Herrin (1979).
Bush-Joseph et al. (1988) also produced a dynamic model to investigate the rate of lifting at either normal (1.1-1.2 m/s), fast (1.7 m/s) and slow (0.8-0.9 m/s) speeds. The model predicted a linear increase in moment with lifting speed. The authors therefore concluded that lifting with excessive speed, such as jerking techniques, should be avoided. The shear and compression components are therefore more sensitive to lifting technique when the effects of acceleration and inertia are included.

Dynamic models have also predicted compression at the inter vertebral joint to be much greater than those predicted by static models. In particular, McGill and Norman (1985) developed a biomechanical model to compare predictions for forces and moments on the L4/L5 joint for static and dynamic tasks. This task was typical of that performed by workers in a metals industry, and involved lifting a sheet metal load of 18Kg from the back of a table, approximately 0.83m away and between mid-chest and waist level and bringing it to rest against the workers abdomen.

The model developed was used to determine the joint reaction forces and moments in three types of analyses. For a dynamic analysis linear and angular accelerations were included. For a static analysis, zero acceleration was assumed. A third quasi-dynamic analysis was also performed which included the force time history for the load in the hands for the dynamic situation, but did not include inertial and acceleration forces. Predictions for the static model showed the peak flexion moment to be approximately 84% of that for the dynamic model, while the quasi dynamic model predicted moments 25% greater than those for the dynamic model. Thus for the dynamic situation considerably higher muscle and inter vertebral joint compression forces were predicted. The authors also suggested the quasi dynamic model avoids the costs associated with dynamic recordings, and could be used as a conservative indicator of the risk of injury.
Estimation of the force generating capacity of each muscle can be calculated as the product of its cross sectional area, and a force coefficient which relates the force to the physiological force producing capacity of the muscle (Rab, 1977). Based on this method of estimating muscle forces, McGill and Norman (1987) produced a model which considered a more detailed representation of the sacrospinalis muscle. The muscle was defined to consist of the lumbar portions of iliocostalis lumborum (IL) and longissimus thoracis (LT) laminae, arising from the iliac crest and each inserting on a different lumbar vertebra. The authors reported the moment arm for each of these laminae was different, with that of the L1 and L2 laminae (which shared a common pathway) being as large as 10 cm. The model was therefore used to compare the effects of a more anatomically detailed muscle force with the predictions of previous models which had assumed a moment arm of 5 cm. Also, a series of parameter studies was performed in which values for muscle cross sectional areas, force coefficient and angle of orientation were varied. The values tested were within ranges reported in the literature, and included those used in previous models.

The flexion moments predicted for sagittally symmetric lifting activities using a previous dynamic model (McGill and Norman, 1985) were used as input to the model. The equivalent extensor moment required for equilibrium for each case was then partitioned among the muscle forces. The contribution of the L3 and L4 sacrospinalis laminae was considered first. The remaining moment was then assigned to the common laminae of the L1 and L2 sacrospinalis, and to the tendons of the thoracic parts of longissimus thoracis and iliocostalis lumborum muscles.
For the cases tested, the greatest reduction in muscle force required was predicted when the LT and IL tendons acted with a moment arm 10 cm posterior to the L4/L5 disc. This was a reduction of 35% in comparison with predictions made when using a single muscle equivalent with a moment arm of 5 cm. The sacrospinalis laminae were also predicted to reduce the anterior shear acting on the disc.

It can thus be seen that the anatomic detail of the models influences the forces and moments they can predict. However, an increase in the number of muscle forces in a model results in an indeterminate system in which the number of unknown forces (muscle and joint reactions) exceeds the number of equilibrium equations to be solved.

2.5.4.3. Three Dimensional Models

A model to determine equilibrium of the spine at a certain level by a number of muscle forces was first reported by Schultz and Andersson (1981). Forces acting on an imaginary cutting plane across the trunk at the L3 level were considered. The moments generated by body weight and external loads above this cutting plane about an arbitrary point were first determined. For equilibrium these moments were required to be balanced by an extensor moment generated by the internal forces. This included unknown muscle forces, the force due to intra abdominal pressure and joint reaction forces.

Five pairs (right and left) of muscle forces were considered which resulted in an indeterminate system of forces to be solved for. This included erector spinae, rectus abdominis, internal and external obliques, and latissimus dorsi. Their action is illustrated in Figure 2-13.
Figure 2-13: A general model for internal force estimation. C, Sa and Sr are the compression, anterior shear and right lateral shear components of the joint reaction considered at the L3 level. Rr and RL are the right and left forces of the muscle rectus abdominis. L represents the force of latissimus dorsi, E erector spinae, X the external oblique and I the internal oblique. P represents the force due to intra abdominal pressure.

(Taken from Schultz and Andersson, 1981.)
Several models including a number of muscle forces have been developed based upon this equilibrium concept. This includes a model developed for the cervical region of the spine (Moroney et al., 1988). However, their method of determination of the muscle forces has varied. Generally the techniques used to determine the muscle activity include:

i) Optimisation
ii) EMG techniques

2.5.4.3.1. Optimisation Assisted Models

In the study of Schultz and Andersson (1981), the forces associated with nonsymmetric lifting situation were solved for using a linear programming technique. A solution for the muscles forces was sought for in which the objective function minimised compression at the inter vertebral joint, but which also satisfied some constraints. These were that muscle forces were non negative (a physiologic impossibility) and that muscle contraction intensity did not exceed a value of 100N/cm², so that the muscle force was within physiological limits. This solution favoured the latissimus dorsi to be active due to its large moment arm.

Marras and Reilly (1988) used the linear programming method to predict the muscle forces and compression at the L3 inter vertebral joint associated with a quasi-static situation. Unlike static situations, this was required to include muscle synergistic activity. Experimental data also collected showed some non linear relationships between muscle activation and trunk velocity and trunk angle, which the linear optimisation procedure could not account for. The authors concluded that linear optimisation is not a suitable means for assessing trunk motion.
Other models have been based upon double linear optimisation (Bean and Chaffin, 1988) and non linear optimisation criteria (Han et al, 1991). These involve more complicated conditions which must satisfy both low muscle intensity and low compression at the inter vertebral joint. A comparison of these techniques was performed by Han et al. (1991). The linear optimisation functions were found to predict a solution which favoured muscles with the greatest mechanical advantage. Muscle coactivity could therefore not be accounted for. In contrast, the double linear optimisation approach gave a solution in which all muscles were active, but the resulting compressive force on the discs was very large. The non linear optimisation function predicted several solutions, which satisfied the constraints representing different possible combinations of muscle recruitment patterns. The predicted compression at the L4 level varied with each solution. However, while the solution which predicts the lowest compression could be identified as the optimal one, the question as to whether this muscle strategy represents that of a living subject remains.
2.5.4.3.2. Electromyography Assisted Models

Electrodes placed over the surface of the back at the anatomical location of the desired muscle groups provide myoelectric detail about the activity of the muscle. The greater the signal, the greater the activity of a particular muscle group. Although the precise relationship between this activity and the actual muscle forces is unknown, the total force required to be produced by the muscles to ensure equilibrium in a lifting action can be partitioned among the individual muscle groups according to their relative activation level. This method has further been refined by including muscle force-length and force-velocity relationships during dynamic lifting (McGill and Norman, 1988a). A particular advantage is that this includes consideration of the difference between right and left sides of the muscle, and also the activity of the antagonistic muscles and synergistic muscles which optimisation procedures cannot account for. This technique also provides data for muscle activity at any stage of a lifting activity (Reilly and Marras, 1989).

Marras and Sommerich (1992 I and II) developed a model of the lumbar spine to investigate the role of muscle coactivation during trunk motion, which few models had previously considered. Subjects performed various isokinetic trunk exertions. The weight, height and torso depth and breadth of each subject at the L5 level were used to estimate the moment generated by the upper body about the L5 inter vertebral joint. The extensor moment required for equilibrium was partitioned among the muscles based upon the relative magnitude of their cross sectional areas and the relative activation using EMG signals. The cross sectional area and moment arms for the trunk muscles were also estimated based on these gross trunk dimensions. These were the internal and external obliques, rectus abdominis, erector spinae and latissimus dorsi.
Force and moment equations for equilibrium were written in order to predict the resulting compressive stress on the spine. These accounted for trunk flexion angle, angular velocity and output torque and were solved repeatedly over the duration of the lift. Experimental data collected for subjects during the tasks included electromyographic data showing the muscle activity throughout the lift, and the level of torque and velocity of the trunk monitored by an isokinetic dynamometer. A comparison between the model predicted torque and measured torque showed good agreement. The model thus allowed trunk velocity, trunk asymmetry and load level which typically vary in the workplace to be investigated.

A most recent model by Granata and Marras (1995) was used to quantify the effects of muscle synergists and antagonists. Subjects performed isokinetic lifts at 0, 30 60 and 90° per second, and free dynamic lifts (fast, medium or slow rates). This involved subjects extending from a trunk flexion angle of 45° to an upright posture. Three dimensional dynamic trunk motions were measured with a lumbar motion monitor (Marras et al., 1992). Three dimensional muscle forces were predicted from EMG data, and their cross sectional areas and dynamic lines of action. Length-strength and force-velocity relations were considered.

An iterative procedure, adopting the EMG activity of one muscle pair at a time, allowed the inclusion of coactive muscles to be quantified. Initially the left and right erector spinae muscle forces were assumed to generate the entire extension moment required for equilibrium. These were considered to be the prime movers for the flexion activity. Additional muscle pairs were then included in turn, until a maximum of five muscle force pairs were predicted. This allowed the effects of muscle coactivity to be studied.
The predicted muscle activity and the resulting spinal compression was influenced by the magnitude of the load lifted and the speed of lift. This agreed with findings previously reported in the literature. Comparison of the predicted compression resulting from the single muscle group with that of the five muscle pairs showed the significance of muscle coactivity. Exclusion of muscle coactivity of trunk flexor muscles during a lifting action was found to underestimate the forces required in the extensor muscles, and the resulting compression and shear on the spine by up to 45% and 70% respectively. An increase in compression on the spine has also been predicted with antagonistic activity during lateral bending (McGill, 1992).

In several EMG assisted models, the predicted muscle forces using EMG data are multiplied by a common gain value so that they agree with force and moment equilibrium equations.

A three dimensional model of the lumbar spine was developed by Guzik et al. (1996) which could predict subject specific and task specific muscle gain parameters. Subjects performed a series of isometric trunk strength tests, for flexion, extension, and right and left lateral flexion. For each task, a dynamometer was used to measure the isometric trunk moment generated. The theoretical moment required to be generated by the muscles for equilibrium was predicted by summing the moment generated by each muscle force. The muscle moments were defined as the product of the cross sectional are and moment arm, and all muscles were assumed active. The gain value was then defined as the measured moment expressed as a fraction of the theoretical moment.
The muscle gain values were greatest for flexion and extension and least for lateral flexion, thereby being significantly different for each task, and were also different for each subject. The gain values for extension were validated with values available in the literature and a good agreement found. However, data was not available for validation of the other tasks. Thus unlike previous studies which had assumed a single gain value for all subjects and all tasks, the current model predicted the necessity for different muscle gain parameters, and was the first of its kind to consider muscle gains for all these activities.

However, the study predicted muscle cross sectional areas from MRI scans without correcting for fibre angle. In addition, all muscles were assumed to be 100% active for each task, but myoelectric measurements were not taken to validate this muscle activity. These two factors may significantly affect the predicted muscle moments, and therefore the gain values. Also, muscle coactivity was not considered.

EMG data has also been used as a source of input to neural network models which can be trained to predict muscle recruitment patterns for given loading conditions (Nussbaum et al., 1995), and also to stochastic models which derive probability density functions from which muscle activity for a given task can be predicted (Mirka and Marras, 1993).
A combination of EMG and optimisation procedures has been used in a model by Cholewicki and McGill (1995). This involves:

i) EMG data was used to partition the required extension moment among the muscle forces according to their level of activation. This was modified for the muscle force-length and force-velocity relationships. In total 50 muscle fascicles were considered. However, only 12 muscle sites could be monitored by surface electrodes with the remaining muscle fascicles grouped assuming the same activation pattern. The authors felt this to be an anatomical improvement over a model which included only 12 muscles.

A common gain value was then calculated and multiplied with the predicted muscle forces so that the best fit between EMG predicted forces and the force and moment equations was found.

ii) An optimisation procedure sought to predict the muscle forces so that the least possible gain value was applied to the individual muscle forces whilst satisfying the requirement to balance moments about the L4/L5 joint in all three planes. In effect this involved minimisation of the square of the difference between predicted EMG forces and those required to satisfy equilibrium, and ensured the gains required were distributed among all muscles rather than just a few.

The hybrid approach showed the advantages of these models to be effectively combined. The model gave an exact match between externally measured and predicted moments, and showed variation in muscle recruitment patterns unique to the subjects and for each trial shown in the EMG study. This therefore accounted for muscle coactivation.
2.5.4.4. Ligaments and Intra Abdominal Pressure

In several models, the force due to intra abdominal pressure has been included to assist the muscles in generating an extensor moment, and to relieve the resultant compression on the spine (Figure 2-14).

![Diagram showing forces](image)

Figure 2-14: Force due to intra abdominal pressure (IAP) acting to relieve the inter vertebral joint compression (C) and to assist in generating an extension moment. (S=shear, L=load, W=body weight, M=Muscle force.)

McGill and Norman (1988a) developed a model to investigate the role of IAP during lifting activities. The force and moment due to IAP was estimated using the various algorithms employed in previous models. The abdominal muscle forces needed to generate an increase in IAP were monitored using EMG techniques and a relationship between abdominal muscle activity and an increase in IAP was found. However, this increase in IAP was accompanied by an increase in predicted compression at the L4/L5 joint. It was calculated the extensor moment produced by the IAP was not more than 5% of the total peak moment required for any of the algorithms used to estimate IAP. Also, the flexion moment generated by the abdominal muscle from their contraction was predicted to equal or exceed that produced by the IAP. Similarly the compressive forces resulting on the spine due to abdominal muscle contraction were predicted to counteract the tensile forces produced by the IAP. The authors therefore concluded the abdominal muscle activity required to generate an increase in IAP counteracts the beneficial effects which IAP was proposed to have in previous models.
In addition, the action of the ligaments in producing extension moments has been investigated. The mechanical advantage of the ligaments having larger moment arms than muscles means the efficiency of such a system would beneficial for supplementing the muscles in generating an extensor moment, both by reducing the muscle forces required, and the consequent compression on the spine. However, the collagen fibres within the ligaments are passive tissue, which requires the ligaments to be stretched before a tensile force can be created (Floyd and Silver, 1955).

A recent prediction of ligament and muscle forces during extreme weight lifting using EMG and video fluoroscopy was made by Cholewicki and McGill (1992). This non-invasive method was a valuable means of determining vertebral displacements throughout a lifting activity, from which the ligament points of attachment, and the consequent change in length of the ligaments could be calculated. They predicted the contribution of these ligaments to be insufficient for the ligament to have a major role in resisting forward flexion.

The lumbo dorsal fascia is reported to have the largest moment of all extensor tissues and therefore has the capacity to exert large extensor moments with minimal compressive penalty (McGill et al., 1988b). Gracovetsky et al. (1977) initially proposed when the muscle power is insufficient to balance the external moments acting at the inter vertebral joint, the muscles act to readjust the spinal geometry so as to activate the lumbo-dorsal fascia and other ligaments sooner than they would be activated passively by an increase in curvature of the spine in flexion. The muscle action was considered to act as a pressure distributed along the fascia, which effectively pushes the fascia outwards away from the spine (Figure 2-15), and stretches it thereby generating a large tensile force. This also increases the moment arm of the ligament tensile force, thereby increasing its moment generating capacity.
A later proposal by Gracovetsky et al. (1981) was that the abdominal muscles contract and exert a lateral pull on the lumbo dorsal fascia through their anatomical attachments to it. A mechanism is described whereby this lateral pull is converted to longitudinal tension, and the resulting tension may be controlled by the degree of firing of the muscles, or the degree of flexion of the spine as mentioned previously. However, the effectiveness of this mechanism requires the muscle fibres to remain in the same place, which in turn depends upon the abdominal cavity being pressurised to a sufficient level.
The above theories could not be validated experimentally. However, following these proposals, the role of the lumbo dorsal (or thoraco-lumbar) fascia has been considered in depth through detailed anatomic investigations (Bogduk and Macintosh, 1984 and Macintosh et al., 1987). Systematic resection revealed that in the posterior layer of its fibres, superficial and deep fibres ran caudomedially and caudolaterally respectively forming a series of overlapping triangles (Bogduk and Macintosh, 1984). This triangular arrangement allows the lateral forces applied at the apex (from the contraction of attached abdominal muscles) to be transmitted along the fibres towards the midline and transformed into resultant longitudinal forces (Figure 2-16) (Macintosh et al, 1987).

Figure 2-16: Lateral tension applied to the posterior layer of the thoraco-lumbar fascia will be distributed in a triangular fashion towards the midline. (Taken from Macintosh et al. 1987.)

The resultant forces act vertically on the spinous process, tending to pull them closer together, thus producing an anti flexion force. However, the limited attachments to the abdominal muscles, and the limited firing capacity of these muscles means this mechanism cannot generate the sufficient anti flexion force for it to be a dominant extensor-moment generating mechanism.
Quantification of the force producing capacity of this structure has been performed by McGill and Norman (1988b). EMG data and joint marker positions were recorded for dynamic lifting activities for various loads in the sagittal plane. A previous model (McGill and Norman, 1986) was used to partition restorative moments among the ligaments, disc and muscle forces. The abdominal forces of latissimus dorsi, internal obliques and transverse abdominis predicted by the model were used to estimate the potential forces generated by the lumbodorsal fascia. However, EMG activity for these muscles was low, and the estimated moment for the LDF was not greater than 4.1% of the total restorative moment required. The muscles were therefore considered the dominant means of producing the required extensor moment.

2.5.4.5. Multi-Segmental Models

More recent multi-segment models have included considerably detailed descriptions of lumbar muscle forces. Detailed systematic resection of individual muscle fascicles in cadaver specimens have been performed for the lumbar erector spinae (Macintosh and Bogduk, 1987 and 1991), multifidus (Macintosh and Bogduk, 1986) and psoas (Bogduk et al., 1992b). Force estimation from computer tomography (CT) scans (McGill et al., 1988) and magnetic resonance images (MRI) (McGill et al., 1993) have also been performed. In particular, the latter study was the first of its kind to provide the data for the muscles acting at both thoracic and lumbar regions. These studies have provided the anatomical detail for the development of multi-segmental models analysing the effects of muscles not only at the level to which they are attached, but also at the levels which they span.
Bogduk et al. (1992a) developed a model to predict the extensor moment generated by the action of every fascicle of the major lumbar back muscles. Based on the previous anatomic studies the three dimensional orientation of the lumbar fascicles was determined by plotting their attachment points onto radiographs of subjects in both the lateral and frontal views. Figure 2-17 attempts to illustrate the detail involved.

![Figure 2-17: Illustrating the points of attachment of the thoracic fibres of longissimus thoracis in the postero-anterior and lateral views. (Taken from Bogduk et al., 1992a.)](image)
At each vertebra level in the lumbar region the forces resulting from muscle contraction for every fascicle acting at that level were summed. This included muscles attached to, and spanning that level. In this way the total extensor moment produced by the back muscles was calculated. Many previous models had assumed a force coefficient with a value of 10 N/cm² (taken from the literature), relating the actual force produced to its physiological capacity. However, the validity of this value is unknown. A particular feature of this model was that values for the muscle forces were expressed as multiples of the force coefficient k, represented by a parameter. Since these values were unknown at the time, this maintained the model’s validity whilst providing the framework for values to be substituted at a later time when known. However, this model did not include consideration of the external moments generated by body weight and external forces, but rather was to indicate the relative force and moment generating capacity of the muscles.

A later model by Stokes and Gardner Morse (1995) investigated the transmission of both forces and moments through the motion segments. This included data estimated for both multi-joint and antagonistic muscle action from cross sectional area and moment arm values obtained from the literature.
Multi-segmental models have also allowed stability of the lumbar spine to be investigated. This is the ability of the spine to maintain or return to its original configuration under the action of an applied load. Bergmark (1989) reported “to avoid collapse of a mechanical system satisfaction of the conditions of equilibrium of the system are necessary but not sufficient. In addition, mechanical stability must be maintained”. Single level models do therefore not investigate the mechanical state of the lumbar spine in full. A three dimensional model of the lumbar spine developed by Bergmark was the first to investigate the stability of the lumbar spine. The spine was required to have a critical stiffness above which it would be stable and below which it would be unstable. This stiffness was provided by the muscles and ligaments of the system. Stable equilibrium was therefore satisfied if the muscle and ligament stiffness was greater than the critical stiffness for the given vertebral configuration. This corresponded to a system with the minimum potential energy.

The spinal muscles were classified into two groups:

i) global muscles which attached to the ribcage and the pelvis. The role of these was to balance the external moments and satisfy equilibrium.

ii) local muscles which attached to the vertebrae (excluding the psoas muscle). These were to control the curvature and provide sagittal and lateral stiffness, and maintain mechanical stability of the lumbar spine. These included interspinalis, intertransverse, multifidus and quadratus lumborum, constituting forty individual muscle fibres.

Hence unlike previous single models which had only considered the spinal muscles with large moment degenerating capacity, Bergmark (1989) included a role for the smaller intersegmental muscle of the spine.
Fixed values for the forces generated by the quadratus lumborum and intertransverse muscles for each run were assumed, whilst forces due to interspinalis and multifidus were distributed according to the approximate relative size of their cross sectional area. This allowed the spinal system to be reduced to a determinate one. The stiffness properties of the muscles were assumed to be proportional to the force generated, while passive stiffness properties of the ligaments were determined according to their change in length corresponding to vertebral configuration.

Thus for a given lumbar configuration, muscle and ligament forces which would satisfy equilibrium and provide the corresponding stiffness greater than the critical stiffness of the system were sought.

A model by Crisco and Panjabi (1991) was developed to investigate the roles of both the deep and superficial lumbar spine muscles. Based upon the minimum potential energy method of Bergmark (1989), the minimal muscular stiffness (defined as the critical muscular stiffness) required to stabilise the spine under these applied loads was determined. The spine would buckle in the lateral plane if muscular stiffness was less than the critical stiffness value. Various muscle architectures were investigated, including the effects of intersegmental muscles, muscles spanning two segments, muscles spanning all five lumbar vertebrae, and a combination of intersegmental and multi-segmental muscles. The authors found the model with only intersegmental muscles was least stable, whereas as the number of segments spanned by multi-segmental muscles increased, the efficiency of stabilisation increased (by 90%). In particular, the model with a combination of the multi-segment muscles (spanning all five vertebrae) and intersegmental muscles showed a 7.5 % increase in efficiency above that of the multi-segmental muscle model alone. This was therefore predicted to be the most efficient muscle architecture.
Cholewicki and McGill (1996) later investigated the stability of the lumbar spine for subjects performing various three dimensional tasks. These included lifting a weight off the floor with one or two hands, holding a load in the upright posture, pushing and pulling with one hand, and performing an upwards sweeping motion with a twist. These therefore covered the full range of movement possible by the spine.

It was assumed each inter vertebral joint contributed a constant proportion of the total movement of the lumbar spine in all three directions. By determining the angle of rotation between the ribcage and the pelvis using an electromagnetic measurement system, the configuration of the spine for each stage of the task could be reconstructed.

Ligament and disc passive stiffness properties were estimated based upon the change in length of their collagen tissue which accompanied lumbar spine motion. The remaining moment required for equilibrium of the external forces and moments was then partitioned among the muscle forces. This model included a total of 90 muscle fascicles. The force generating capacity of each was estimated from cross sectional areas, whilst the corresponding stiffness was estimated using a cross bridge bond distribution moment model described in the literature. The combined EMG and optimisation approach previously reported (1995) was used to determine the relative activation of the muscle groups.
By considering the configuration of the spine at various time intervals during a task, the necessary muscle and ligament forces for equilibrium were determined. The corresponding stiffness generated by these elements was also determined, and was used to establish an index of stability for the given configuration. The main finding was that tasks which required high muscle activity resulted in greater stability than less demanding tasks in which muscle activity and stability was low. The stability of the spine therefore varies with the posture and muscle forces associated with a given task.

With developing computer technology, finite element analyses which were once restricted to single motion segments due to the complexity involved have now been extended to include a number of motion segments, and indeed some cases, the whole lumbar spine. Shirazi Adl and Parnianpour (1993) developed a finite element model of the whole lumbar spine in order to investigate the stability of the spine under certain loading conditions. Due to the line of gravity of the trunk passing anterior to the spine, body weight generates an axial compressive force and a flexion moment on the spine. In addition, intra-abdominal pressure and intramuscular pressure (which increase with abdominal muscle contraction) were thought to provide lateral pressure and help stabilize the spine. The effects of axial compression, flexion moment and a horizontal support were therefore investigated. The model formulation accounted for material and geometric nonlinearities. Loading conditions were applied incrementally:

i) axial compression alone
ii) axial compression + horizontal support
iii) axial compression + flexion moment
iv) axial compression + flexion moment + horizontal support
Boundary conditions for the L1 vertebral body were also varied so that it was completely free, constrained for sagittal translation, and constrained for sagittal and lateral translation. The presence of a flexion moment increased the stiffness of the spine in the sagittal plane, and decreased the displacements, thus stabilising the lumbar spine. The horizontal support simulating intramuscular pressure and intra abdominal pressure had a similar effect in the coronal plane.

Also, the model predictions showed under axial compression the lumbar spine underwent large sagittal displacements, and smaller lateral displacements, and resisted loads up to 200N. Beyond 200N buckling did not occur as for a simple column due to the complex system of the lumbar spine, although a softening effect did occur with diminishing stiffness in the coronal plane. This was defined as instability, and was only observed due to the incremental nature of loading. The authors suggested lumbar lordosis and body weight act together so that the maximum axial load can be withstood, with minimum muscle activity, thereby reducing the energy cost for stability. Thus unlike previous authors who had represented the spine as a column and defined instability using buckling theory, the authors showed the spine to be a more complex system which exhibited a point beyond which stiffness decreased in the coronal plane, but buckling did not occur. This model was later developed into a thoraco-lumbar spine (Shirazi Adl and Parnianpour, 1996) and is discussed in the appropriate section.
An alternate model to investigate equilibrium and stability of the spine was proposed by Aspden (1987). This involves representing the spine as an arch structure which is capable of balancing moments internally. Muscle forces were proposed to act along the length of the spinal arch providing the necessary compression to stiffen and strengthen its structure, allowing it to withstand the loads applied. The degree of force required depends upon spinal curvature; less muscle force was required with a greater curvature. The curvature of the spine was therefore thought to have a load bearing role. The path of the resulting forces within the structure is known as a thrustline. The existence of a thrustline indicates equilibrium, and its coincidence with the configuration of the vertebral bodies so that forces are transmitted between adjacent vertebrae indicates stability. In addition, a new role for intra abdominal pressure was proposed whereby the force due to IAP provided stability to the lumbar spine by acting perpendicular to the anterior surface of the lumbar vertebral bodies. A model of the thoraco-lumbar spine was developed, and a position of 30 degrees of forward flexion considered (Figure 2-18).

Although this model includes thoracic and lumbar vertebrae, the model predictions for the L5/S1 joint were compared with a lever model (Morris et al., 1961). For this reason the model has been considered in the lumbar region section. The loading conditions of the lever model of Morris et al. (1961) were simulated precisely. However, whereas the predicted compression at the L5/S1 joint was 6600N for the lever model, the arch model which included the new role for IAP and curvature predicted 1500N. The model predicted the dynamic curvature may therefore have a role in load bearing.
The same model was then used to investigate combinations of posture and loading which could be recommended as 'safe' to adopt in the workplace (Aspden, 1988). However, in this model the spinal muscles were represented as a collective force at the upper end of the arch. The anatomic detail therefore remains to be developed.

Figure 2-18: Illustrating the loading conditions of the lever model applied to the spine as an arch. W1 and W2 represent forces due to body weight, and F the external load carried. IAP acts over the lumbar region of the spine. (Taken from Aspden, 1988.)
Aspden (1992) has also reported the integrated structure of the spine should be considered as a system with each element being co-ordinated to control the curvature of the spine, thereby maintaining stability and controlling of the dynamic structure. The absence or deficiency of an element at one level may upset the whole balance of the system, not just at the level of the defect. Certainly therefore a more realistic representation of the spine as a complete system, with a complex interaction of muscles, ligaments, abdominal pressure and curvature is very important for future models.

The effects of curvature of the lumbar spine have also been investigated using finite element techniques. Lavaste et al. (1992) performed a morphologic study of forty cadaver lumbar vertebrae and identified six main parameters which could describe the vertebra geometry entirely. These were vertebral body width, depth, height and concavity, and the total height and depth of the vertebra. These are shown Figure 2-19. From these parameters all other component dimensions could be calculated.

Figure 2-19: The main parameters: vertebral body width (A), vertebral body depth (B), vertebral body height (C), vertebral concavity (E), total vertebra height (H) and total vertebra depth (L).

(Taken from Lavaste et al., 1992.)
The position and angle of orientation of each vertebra was described so that the entire lumbar spine, including its lordotic curvature could be reconstructed (Figure 2-20). In this way the geometry of a given lumbar vertebra could be reconstructed realistically using simple parameters. Material properties were also assigned to the various components, so that mechanical investigations could be performed. This included testing of the lumbar spine under applied compression, flexion, torsion and lateral bending. Inclusion of the entire lumbar spine also allowed the effects of the lordotic configuration to be investigated. The authors report the value of their model was the ability to represent the geometry of any given lumbar specimen. An exact comparison between experimentally tested specimens and model predictions could therefore be made.

Figure 2-20: Three-dimensional reconstruction of the lumbar spine.
(Taken from Lavaste et al., 1992.)
A detailed finite element model of the lumbar spine was developed by Shirazi Adl (1994) to investigate the response of the lumbar spine to sagittal and lateral bending moments. The roles of the ligaments, annulus fibres and facets were investigated. The geometry of the model was obtained using CT scans.

Flexion, extension and lateral bending moments were applied to the model, and the corresponding displacements predicted. Cases investigated included a variation in the facet gap limit, which represented the state of degeneration of the articular cartilage in the facets. Removal of a single facet or both at the L4/L5 level was also investigated, and was simulated by removing the possibility of articulation between the two surfaces. The results predicted in extension removal of one or both facets increased flexibility only at the L4/L5 level. In contrast, a larger gap limit representing more effective articulation at the facet joints was predicted to increase the stiffness at all levels of the lumbar spine, and to decrease the disc pressure at all levels. This demonstrated the interactive load bearing roles of the discs and facets at all lumbar levels. The three dimensional lordotic geometry of the lumbar spine was also predicted to result in coupled motions. This illustrates the importance of considering spinal curvature in these models.
2.5.5. UNIT MODELS

A number of models have been developed to investigate in detail the properties of the vertebrae or the inter vertebral disc, or how they function as a unit. These have mainly been based upon finite element techniques which allow the microscopic investigation of structures. In particular, the displacements, strains and stresses can be predicted for each region of the motion segment. The considerable number of models in the lumbar region requires further classification as either vertebra, disc-body unit or motion segment models. The majority of cervical and thoracic unit models are relatively recent, and employ the finite element modelling techniques well developed from earlier lumbar studies. The studies have varied in their representation of the elements and the type of loading applied.

2.5.5.1. Cervical Unit Models.

A three dimensional model of the axis vertebra (C2) was developed by Teo et al. (1994) in order to determine stress distributions under various directions of loading. The geometry for the model was digitised from a cadaver specimen using a coordinate measuring machine. These coordinates were used to generate a finite element mesh (Figure 2-21). It was assumed the vertebra consisted totally of cortical bone, thereby ignoring the cancellous bone, while the discs, transverse ligaments, superior and inferior facets were all modelled by spring elements.
An arbitrary force of 1KN was distributed over the anterior surface of the odontoid process in the antero-posterior direction. Angles of 0, 45 and -45 degrees relative to the vertical for this force were used to represent respectively impact situations of the head on a windscreen when in an upright position, the head on an obstacle when the neck is in extension, and an upward punch incurred in boxing. Regions of high compressive stress were predicted at the posterior surface of the dens, and the inner lateral surface of the superior facets, while maximum tensile stress was found at the junction of the dens and the vertebral body. The results also showed this stress distribution to vary with the direction of loading. The authors reported these regions of high stress corresponded with recorded fracture locations recorded clinically and from and judicial hangings.
Bozic et al. (1994) investigated the burst fracture mechanism of a cervical vertebral body. A three dimensional finite element mesh was constructed using quantitative computer tomography (QCT) techniques from a 66 year old male cadaver specimen. Young's modulus for the bone material was derived by calculating the apparent density for each pixel on the CT scan. Equations relating the elastic modulus and density, and bone strength with density were then used, and values for elastic modulus and strength of each element were then computed by averaging the values for each pixel occupying volume in that element. In this way three dimensional variation in material properties could be accounted for.

In addition, spring elements were attached to the vertebra on its superior and inferior surface representing the presence of inter vertebral discs. Springs located in the position of the nucleus were given stiffness properties four times greater than those peripheral ones representing the annulus fibres. A uniform axial displacement applied to the superior springs provided a varied stress pattern deemed more realistic than one achieved with uniform application of compressive load.

The resulting stress field was three dimensional, indicating the necessity for the three dimensional detail incorporated in the current model. A factor of safety, relating predicted shear stress to shear strength was defined. Model predictions indicated fracture would begin in the central cancellous bone of the vertebral body at loads significantly lower than the equivalent 3400N applied here. This agreed with a previous burst fracture study on a thoracic vertebra (Oxland and Onat, 1993) and with fracture lines seen clinically in burst fractures. The results therefore indicated that high compressive loading may be a cause of burst fractures, and that by avoiding such loading conditions, the devastating effects associated with it may be prevented. However, for further validation, an vitro testing would be required.
A more detailed review of these models was given by Yoganandan et al. (1987). In particular, the lack of information regarding material properties was noted. The authors report models in this area may therefore be useful in performing parametric studies to build a database of the required information, and also to investigate the effects of instrumentation used to correct disorders in this region.

2.5.5.2. Thoracic Unit Models

Oxland and Onat (1993) developed an axi-symmetric model of a T12 vertebral body in order to investigate the mechanism of burst fracture under axial compressive loading. In particular, the effects of disc degeneration on this mechanism were considered. The inter vertebral disc was simulated by a uniform compression acting on the radius of the vertebra. For a normal disc this was 75% of the radius, whereas in degeneration the pressure was assumed to act across the entire radius. To investigate the fracture mechanism the axial compressive load was increased until failure occurred. Failure was defined by a Von Mises stress criteria for each material.

The model predicted under pressure from a normal inter vertebral disc the cancellous bone in the vertebral centrum failed first, due to compressed trabeculae. Following this the superior end plate depressed into the cancellous bone resulting in end plate failure. The cortical bone on the periphery of the vertebral body then failed due to large tensile hoop stress. These findings were consistent with those of previously published experiments.
In contrast, under pressure from a degenerated disc the peripheral cortical bone was predicted to be the first structure to fail due to large axial and hoop stress. The central cancellous bone in the vertebral centrum then failed. The authors therefore suggested that the thoraco-lumbar burst fractures are relatively unlikely in vertebrae next to severely degenerated discs, again supporting previous experimental findings in the literature. They report the value of their model was its ability to predict the multi stages of injury mechanism for burst fractures, by investigating the progression of failure beyond the initial failure point.

2.5.5.3. Lumbar Unit Models

Unit models of the lumbar region of the spine employing finite element techniques are numerous and vary widely in their applications. The previous review by Goel and Gilbertson (1995) has provided a comprehensive review of finite element models and their value in clinical applications. The reader is referred to this paper for further reading. The current review therefore considers a small number of additional models which can provide the overall picture of the development of such models. The detail of these models, and their value can then be appreciated alongside the simpler models of larger sections of the spine which do not employ finite element techniques.
2.5.5.3.1. Vertebrae

Although many models have focused upon the behaviour of the disc due to its critical role in load bearing, the large deformations it may undergo, and the disorders which may result, the vertebra itself is prone to bone disorders, such as osteoporosis and metastatic lesions. Further, under axial compression the vertebra and endplates are likely to fracture before the annulus or disc ruptures, whilst the effects of muscle action have been reported to result in a region of particularly high stress at the pars interarticularis, and at the junction of the lamina with the pedicles (Dietrich and Kurowski, 1985). Models have therefore been produced in order to investigate the stress distribution within the vertebra.

Hakim and King (1979) were the first to produce a three dimensional finite element model of a lumbar vertebra which included the posterior elements. This included bilateral symmetry, and the geometry for half a vertebra was simulated using a semi automatic finite element mesh generation technique. In a later study, Balasubramanian et al. (1979) extended this model (Figure 2-22) to include a whole vertebra, so that the effects of the surgical procedure of laminectomy could be observed. Experimentally measured pressures were determined for combined compressive (2390N) and shear (420N) loads. The resulting pressure was applied in the model at the superior end plate. The results showed that stress was highest at the junction of the vertebral body with the pedicles, and that a partial laminectomy caused the highest stress values. The study was thus valuable to surgeons showing a complete laminectomy is more beneficial than a partial one.
Due to hormonal changes, post menopausal women are particularly at risk from osteoporosis. In this condition the elastic modulus of the vertebrae is reduced leaving them prone to fracture. Conventional quantitative computerised tomography (QCT) scans have been used clinically to predict the fracture threshold of the vertebra based upon the trabecular bone mineral density. However, Faulkner et al. (1991) reported that a scan of a single section does not represent the complete vertebral structure. They therefore produced a three dimensional finite element model which would predict the failure threshold for the vertebra in terms of yield stress. This was defined as the resulting stress beyond which the cortical and trabecular bone could no longer support any load. The conventional QCT scan was used to provide information for the bone density and distribution of a vertebra, and also for its geometry for both normal and osteoporotic female patients. Only the vertebral body was included as these were assumed to be the main load bearing element, and the predominant site for osteoporosis.
Data for the elastic modulus, Poisson ratio and yield point for the trabecular and cortical bone were assigned from the literature based on the average bone mineral density of the specimens and strain rates. Loading was applied incrementally.

The traditional fracture threshold predicted from the CT scans for trabecular bone mineral density was compared with the predicted failure threshold predicted in terms of yield stress. Three dimensional mesh plots were generated and overlaid with the results for displacement and strain contours obtained. This allowed rapid visualisation of the model and localisation of the potential weaknesses in the vertebral structure. It was revealed there was an overlap in BMC and TMD for normal and osteoporotic patients, indicating it is difficult to define a border between normal and osteoporotic bone based on just BMC or TMD alone. In comparison, yield strength values predicted by the model showed minimal overlap between the two categories. The authors therefore reported their model could predict vertebrae with signs of osteoporosis based upon yield strength better than the QCT clinical techniques based upon trabecular mineral density values.

Mizrahi et al. (1992, 1993) also developed mathematical models to investigate the effects of age related osteoporosis on vertebral fractures, and the effect of metastatic lesions in cancer patients on the stress distributions within the vertebral bodies. These disorders could be simulated by altering the material properties of the vertebral cortical and cancellous bone. These have been further reviewed by Goel and Gilbertson (1995) who emphasise the value of these models in investigating clinical disorders.
2.5.5.3.2. Disc_Body Unit Models

Some models were developed to identify important material and geometric properties of the disc and vertebral body (Spilker, 1980 and Suwito, et al., 1992), so that future models can be increased in complexity while including just these main features. These studies have involved some simplifications to allow series of parameter studies to be performed.

The model of Suwito et al. (1992) involved a simple axi-symmetric representation with isotropic material properties. Half a lumbar spine motion segment was represented in this way, excluding consideration of the posterior elements, asymmetry of the vertebral body or disc, and lumbar lordosis.

The nucleus was represented by applying a prescribed pressure to the cartilaginous endplate and disc annulus. Variation of this pressure allowed a wide range of disc conditions such as ageing or denucleation to be studied. For a basic model, these material and geometric properties were assigned from published data available. However, the absolute values were not considered important, as the study was intending to determine the relative effects of each parameter.

Parameter studies were performed to determine the effects of variation in the material properties, including the hard and soft regions of the cancellous bone. Geometric properties included the height of the vertebral centrum, curvature of the endplate, and disc radius. The peripheral curvatures of the vertebra and disc were included as these were hypothesised to have an influence on the stress-strain results.
In this study the annulus was assigned orthotropic material properties, with an elastic modulus describing the stiffness of the fibres in the radial, longitudinal and circumferential directions. The modulus values were also varied so that they were least at the interior of the disc and increased towards its perimeter.

A more detailed representation of the annulus was developed by Shirazi Adl et al. (1984) and has been discussed by Liu and Goel (1987). This involved recognition that the annulus is a non homogeneous composite. The fibres were oriented in a criss cross manner and were arranged in layers, embedded in an amorphous ground substance (Figure 2-23). The fibres and the ground substance were assigned separate material properties.
Shirazi Adl (1989a) later compared the predictions of this model representing the annulus as a non homogeneous composite, with those of previous models as orthotropic homogenous composite. For simplicity symmetry about the vertical axis and about the mid horizontal plane was assumed.

For the nonhomogenous representation of the annulus, a collagen fibre content of 16% of the annulus volume was assumed. The remaining volume was attributed to a matrix of ground substance. The ground substance was modelled by homogenous isotropic material properties. The fibre layers were represented by circumferential membrane elements which resist loads only in the direction of the fibres. Eight fibre membrane layers were represented with properties which varied from the innermost to outermost layer, with the inner layers assigned lower stiffness values. This variation in material properties within the annulus layers was also included in the orthotropic representation.
Material properties were assigned so that the overall behaviour of the annulus, vertebra and nucleus in both models was the same. The same displacement was also prescribed to both models, so that the same overall behaviour for each would be obtained. An updated Lagrangian formulation was used in the analysis to accommodate for both these geometric and material nonlinearities. For both models the predicted displacements and strains were the same. However, due to the different representations of the annulus, the nonhomogenous representation predicted different stress distributions compared with the orthotropic model. However, within the non homogeneous composite, due to the difference in material properties assigned to the fibres and the ground substance, there was a different stress response. This reflected the distinct behaviour of the annulus fibres and ground substance. The study thus showed that while the overall behaviour of the disc may be determined with either a simplified or a complex representation of the annulus, the distribution of stress within the structure may be affected by the composition.

The inclusion of the ground substance thus affects the behaviour of the disc. Determination of the material properties of the ground substance can be achieved experimentally by analysis of the initial linear portion of the obtained load-displacement curve for the disc; the later non linear portion attributable to the behaviour of the collagen fibres (Rao and Dumas, 1991). However, due to its interaction with the collagen fibres, its response to loading cannot be investigated experimentally, demonstrating how models can investigate the inter vertebral disc to a level of detail greater than that possible through experimental observation.
Some recent studies have also investigated the response of the disc under prolonged loading. Natali (1991) investigated the effects of nucleus degeneration in three dimensional by modelling it as a hyper elastic material. This allowed geometric and material non linearities, creep and relaxation to be accounted for. The internal energy of the nucleus material was defined using non linear equations, from which the variation of internal energy, the displacements, and ultimately a strain energy potential was defined. This was then used to evaluate axial displacements, disc bulge, bony endplate bulging, and intradiscal pressure as a function of the applied axial compression. A compressibility coefficient within the strain energy potential formula was used to define the state of compressibility of the nucleus, and hence the state of degeneration. A value of zero simulated a healthy nucleus with a state of incompressibility.

A poroelastic approach (Simon et al., 1985) has been used to investigate the effects of nuclear degeneration under transient and steady state loading, and the effects of osmotic pressure. These have been discussed more fully by Goel and Gilbertson (1995). Ideally, these models may be indicative of the behaviour of the inter vertebral disc, whilst in any one posture for a prolonged period of time. However, such models have yet to be applied to the analysis of the loads involved with sedentary office tasks.
For workers exposed to levels of vibration associated with vehicle operation, a high incidence of low back pain has been reported. Kasra et al. (1992) performed a dynamic analysis on the disc-body unit in order to investigate the effects of various levels of vibration on the spine. Both experimental and finite element investigations were performed. The finite element models were both three dimensional and axi-symmetric models, thus involving linear and non linear representations. Testing cases included a compressive preload, to represent the influence of body weight and external loads carried which had not previously been considered. The output of the study was considered in terms of compliance as a function of frequency. In addition, the influence of facectomy was performed experimentally, and the effects of nucleus fluid loss and nucleotomy were predicted in the models.

2.5.5.3.3. Motion Segments

Representation of two adjacent vertebrae and their interconnecting disc allows the interaction between the inter vertebral disc, the articulating facets and the ligaments to be investigated. This is known to be significant during flexion, extension, lateral bending and torsion. Further, the facets are known to be susceptible to degenerative processes and injuries and possible sources of low back pain (Shirazi Adl et al., 1986). Some models were therefore produced in order to investigate the load bearing paths of the facets and ligaments (Shirazi Adl et al., 1984, 1986 and Ueno and Liu, 1987).
Most significantly, Shirazi Adl et al. (1986) was the first to represent the facet articulation as a non-linear contact problem, and to include the posterior ligaments in their model of a motion segment. Studies prior to this had represented the facets using linear elements with a stiffness value which remained constant. An updated Lagrangian approach was used which allowed the joint configuration to be analysed and updated at each incremental loading step. The changing state of the facet articulation, either in contact or separating according to the loads applied could thus be treated as a 'moving contact problem'. Contact between the facets was defined when the distance between their curved surface decreased to less than a specified value ('gap limit'), and the corresponding stiffness value and resistance to loading was increased. In a later study (Shirazi Adl and Drouin, 1987) the finite element code was modified slightly to yield both the magnitude and direction of the facet forces. Figure 2-24 illustrates the detail of the articulating facet surfaces modelled.
The ligaments were also included, represented by axial elements oriented along the direction of their fibres. Figure 2-25 illustrates their attachments to the motion segment.

![Figure 2-25: Illustrating the representation of ligaments by uni-axial elements oriented along the fibre direction. (Taken from Shirazi Adl and Drouin, 1987.)](image)

A more recent study by Shirazi Adl (1991) investigated the role of facets under combined loading using the previous non-linear three-dimensional model (Shirazi Adl et al., 1986). The behaviour of the facets was investigated under axial torque and lateral bending, and axial torque and lateral bending with 1000N axial compressive preload. In addition, a case of symmetric heavy lifting was simulated with the relative magnitude of anterior shear force on the facets as either 10% or 20% of the applied compression. A case of nonsymmetric lifting was also simulated by applying axial torque and lateral bending. The loads were applied incrementally with the final values for these cases and the resulting rotations chosen to agree with loads and displacements associated with lifting while in full flexion reported in the literature.
The axial, sagittal and lateral components of the predicted facet contact force were computed by scalar addition of the predicted individual elements at the various contact regions. The resultant force at the facets was therefore determined by vectorial summation of the three components.

The model predicted three distinct sets of contact areas for the facets under different forms of loading. In extension, the sagittal and axial components of force were larger than the lateral ones due to the vertical inclination and coronal orientation of the facets. However, in flexion the contact loads were almost horizontal, with negligible axial component. The sagittal component that resists forward translation of the upper vertebra was predicted to be larger than the lateral one as expected. In contrast, in torsion the vertical and sagittal orientations of the surfaces produced contact loads that were nearly laterally oriented, with negligible components in the sagittal and axial directions. The facet contact loads were thus significantly determined by facet geometry and loading.

In addition, the model was effective in demonstrating the magnitude of the facet contact pressure, and the contact areas. The authors report that further refinement in the number of nodes used to model the facet surfaces would give an even more accurate prediction.
Shirazi Adl (1989b) also developed a model to investigate the degenerative process of the inter vertebral disc under loading associated with heavy symmetric and non symmetric lifting activities. In particular, an association between frequent heavy lifting and a high prevalence of disc prolapse had been reported. The disc annulus was represented using the non linear inhomogeneous composite of collagen fibres embedded in a matrix of ground substance and the facet articulation was represented using the moving non linear frictionless contact problem previously defined. Ligaments were also included as a set of uni-axial elements with non linear material properties. The finite element mesh generated was for an L2/L3 motion segment, with symmetry assumed about the sagittal plane.

The model was tested for single applied loads (sagittal moments, lateral moments, axial torque and compression) and then for combined loading. The maximum applied loads were in a similar range to those reported to occur during heavy lifting tasks. One particular case investigated involving a combination of axial torque and lateral bending simulated lifting whilst bending to one side. For this case a failure analysis was performed. Collagen fibres strain values were set as 14% yield and 16% ultimate tensile strains. Fibre elements with strains exceeding these values were removed at each stage of loading. In this way failure progression of the fibres could be investigated. Degeneration of the disc was also simulated through loss of intradiscal pressure.
The model predicted under heavy non symmetric lifting the innermost fibre layer of the annulus was likely to rupture at the postero-lateral location. Under this loading, the nuclear pressure was predicted to be large, so that the nucleus would be likely to penetrate into the fissure and result in disc prolapse. With loss of nucleus fluid and therefore reduced pressure, the model predicted the likelihood of prolapse would be reduced. The study was validated by comparison with experimental studies which had examined the surface strains of the annulus.

A three dimensional finite element model of an L2/L3 motion segment was also recently developed by Lu et al. (1996) in order to investigate the mechanism of disc prolapse. The annulus fibres, and the ligaments of the motion segment were modelled by two dimensional cable elements which could sustain tensile stress only. The fibres constituted 5% of the annulus at the innermost layer, and 23% in the outermost, representing a gradual change in collagen content from outer to inner layers. The nucleus was modelled as an incompressible fluid, represented by three dimensional hydrostatic fluid elements, while the articulating facets were represented by sliding contact elements.

Material properties were assigned to each element based on previous studies. All were assumed linear. However, viscoelastic properties were also assigned to the annulus fibres and ligaments. Loads were therefore applied incrementally so that the effects of viscoelastic properties could be investigated. These loads included pure axial compression, and axial compression combined with 7 degrees flexion and 2 degrees axial rotation. This latter case was also tested with 10% fluid removed from the nucleus in order to investigate the effects of diurnal fluid loss from the nucleus.
Geometric nonlinearity (large deformation theory) was used to account for the relatively large deformations that occur in the disc under loads. At each increment of load, the tensile stress in the annulus was compared with a preset failure stress of 50MPa. Beyond this level the fibres were assumed to be ruptured, and were therefore removed from the model so that they would no longer contribute to the stress pattern.

The final compressive load was 4000N, representing the compressive strength of a lumbar motion segment, after which the model was assessed for disc damage. The results showed the maximum stress in the annulus always occurred at the inner posterior annulus at the junction of the disc and endplate, supporting previous studies which suggest this is the site of fissure initiation. The combined loading case of compression, flexion and axial rotation predicted failure. However, with fluid loss from the nucleus tensile stress in the annulus decreased and failure did not occur. The model also predicted when a fibre ruptured, the stress was redistributed to the surrounding fibres so that additional load was required to cause further failure. The study therefore indicated that a saturated disc (no fluid loss) typical of the state of the disc after resting, subjected to combined loading of compression, flexion and axial rotation may be causal factors for disc prolapse.
2.5.6. DISCUSSION

In the current review, models have been discussed based on a classification scheme which considers whole, partial or unit models of the spine. The partial and unit models are further considered regarding the region of study; cervical, thoracic or lumbar. This provides some order to the immense number of mathematical models available in the literature. A summary of the models discussed is presented in Table 2-1 (whole spine), Table 2-2 (head-neck), Table 2-3 (thoraco-lumbar), Table 2-4 (lumbar) and Table 2-5 (unit).
Table 2-1: Summary of the main features of the whole spine models discussed in this chapter.

<table>
<thead>
<tr>
<th>Authors</th>
<th>Continuous/Discrete</th>
<th>2D/3D</th>
<th>Focus of Study</th>
<th>Analysis</th>
<th>Loading</th>
<th>Features Included</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cramer et al. (1976)</td>
<td>Continuous</td>
<td>1D</td>
<td>Pilot ejection</td>
<td>Dynamic, Non linear</td>
<td>Acceleration (+Gz) upwards at the base of the spine</td>
<td>Sagittal curvature</td>
</tr>
<tr>
<td>Soetching &amp; Paslay (1973)</td>
<td>Continuous</td>
<td>2D</td>
<td>Vehicle collision</td>
<td>Dynamic, Non linear</td>
<td>Deceleration from an initial velocity V, in sagittal plane</td>
<td>Passive muscle forces</td>
</tr>
<tr>
<td>Lindbeck (1987)</td>
<td>Continuous</td>
<td>2D</td>
<td>Flexural rigidity</td>
<td>Non linear</td>
<td>Lateral bending</td>
<td></td>
</tr>
<tr>
<td>Örne &amp; Liu (1971)</td>
<td>Discrete</td>
<td>2D</td>
<td>Pilot ejection</td>
<td>Non linear</td>
<td>Acceleration (+Gz) upwards at the base of the spine</td>
<td>Viscoelastic disc</td>
</tr>
<tr>
<td>Belytschko et al. (1978)</td>
<td>Discrete</td>
<td>3D</td>
<td>Pilot ejection</td>
<td>Non linear</td>
<td>Acceleration (+Gz) at the base of the spine, rate and angle varied</td>
<td>Ribcage, helmet system, parameter study</td>
</tr>
<tr>
<td>Seirig &amp; Arivikar (1975)</td>
<td>Discrete</td>
<td>2D</td>
<td>Postural loading</td>
<td>Linear optimisation</td>
<td>Force due to body weight</td>
<td>Active muscle forces, IAP</td>
</tr>
<tr>
<td>Schultz &amp; Galante (1970)</td>
<td>Discrete</td>
<td>3D</td>
<td>Motion of spine</td>
<td>Geometric</td>
<td>Flexion, extension, lateral bending, axial rotation</td>
<td>No material properties</td>
</tr>
<tr>
<td>Panjabi (1973)</td>
<td>Discrete</td>
<td>3D</td>
<td>Force-deformation relation</td>
<td>Linear</td>
<td>No illustrative spine example given</td>
<td>-</td>
</tr>
</tbody>
</table>

115
Table 2-2: Summary of the main features of the head-neck models discussed in this chapter.

<table>
<thead>
<tr>
<th>Authors</th>
<th>2D/3D</th>
<th>Focus of Study</th>
<th>Loading</th>
<th>Analysis</th>
<th>Muscles</th>
<th>Features</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prasad &amp; King (1974)</td>
<td>2D</td>
<td>Pilot ejection</td>
<td>Acceleration (+Gz) upwards at the base of the spine</td>
<td>Dynamic</td>
<td></td>
<td>First to include facets</td>
</tr>
<tr>
<td>Williams &amp; Belytschko</td>
<td>3D</td>
<td>Vehicle collision</td>
<td>Frontal &amp; lateral impact</td>
<td>Dynamic</td>
<td>Active &amp; passive</td>
<td>Facets as pentahedral element</td>
</tr>
<tr>
<td>(1983)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Merrill et al. (1984)</td>
<td>3D</td>
<td>Vehicle collision</td>
<td>Frontal &amp; lateral impact</td>
<td>Dynamic</td>
<td>7 pairs passive</td>
<td></td>
</tr>
<tr>
<td>Deng &amp; Goldsmith (1987)</td>
<td>3D</td>
<td>Vehicle collision</td>
<td>Frontal &amp; lateral impact</td>
<td>Dynamic</td>
<td>15 pairs passive, curved line of action</td>
<td></td>
</tr>
<tr>
<td>Hoek et al. (1993)</td>
<td>2D</td>
<td>F16 pilots</td>
<td>Gravitational forces</td>
<td>Static, muscles optimised</td>
<td>Active, combinations of three muscle pairs</td>
<td>C3-C7 single link</td>
</tr>
<tr>
<td>Kleinberger (1993)</td>
<td>3D</td>
<td>Vehicle collision</td>
<td>Axial compression and frontal flexion</td>
<td>Finite element</td>
<td></td>
<td>Regular geometry, facets, discs and ligaments included</td>
</tr>
<tr>
<td>Yoganandan et al. (1997)</td>
<td>3D</td>
<td>Parametric study</td>
<td>Axial compression</td>
<td>Finite element</td>
<td></td>
<td>C4,C5,C6 only</td>
</tr>
</tbody>
</table>
Table 2-3: Summary of the main features of the thoraco-lumbar spine models discussed in this chapter.

<table>
<thead>
<tr>
<th>Authors</th>
<th>2D or 3D</th>
<th>Model Representation</th>
<th>Focus of study</th>
<th>Analysis</th>
<th>Features</th>
</tr>
</thead>
<tbody>
<tr>
<td>Belytschko et al. (1973)</td>
<td>3D</td>
<td>Discrete</td>
<td>Study of force-deformation properties</td>
<td>Non linear</td>
<td></td>
</tr>
<tr>
<td>Schultz et al. (1973)</td>
<td>3D</td>
<td>Discrete</td>
<td>Study of force-deformation properties</td>
<td>Non linear</td>
<td>Greater ligament detail</td>
</tr>
<tr>
<td>Takashima et al. (1979)</td>
<td>3D</td>
<td>Discrete</td>
<td>Effects of muscle contractions</td>
<td>Non linear</td>
<td>Active muscle forces</td>
</tr>
<tr>
<td>Schultz et al. (1981)</td>
<td>3D</td>
<td>Discrete</td>
<td>Scoliosis and muscle stimulation</td>
<td>Non linear</td>
<td>Electrical muscle stimulation</td>
</tr>
<tr>
<td>Andriacchi et al. (1974)</td>
<td>3D</td>
<td>Discrete</td>
<td>Ribcage and spinal bending</td>
<td>Non linear</td>
<td>Ribcage</td>
</tr>
<tr>
<td>Wynarsky &amp; Schultz (1991)</td>
<td>3D</td>
<td>Discrete</td>
<td>Scoliosis and bracing techniques</td>
<td>Non linear</td>
<td>Active muscle forces</td>
</tr>
<tr>
<td>Ghista et al. (1988)</td>
<td>2D</td>
<td>Discrete</td>
<td>Scoliosis and surgical correction</td>
<td>Finite element, non linear</td>
<td>Motion segments cylindrical beams</td>
</tr>
<tr>
<td>Patwardhan et al. (1986)</td>
<td>2D</td>
<td>Continuous</td>
<td>Scoliosis and orthotic correction</td>
<td>Non linear</td>
<td>Active muscle forces</td>
</tr>
<tr>
<td>Noone et al. (1991)</td>
<td>2D</td>
<td>Continuous</td>
<td>Scoliosis and muscular dystrophy</td>
<td>Non linear</td>
<td>Active muscle forces</td>
</tr>
<tr>
<td>Stokes &amp; Gardner Morse (1991)</td>
<td>3D</td>
<td>Discrete</td>
<td>Causes of Scoliosis</td>
<td>Finite element</td>
<td>Motion segment coupling, posterior tethering by soft tissues</td>
</tr>
<tr>
<td>Lee et al. (1995)</td>
<td>3D</td>
<td>Discrete</td>
<td>Manipulative therapy</td>
<td>Finite element</td>
<td>Ribcage</td>
</tr>
<tr>
<td>Monheit &amp; Badler (1991)</td>
<td>3D</td>
<td>Discrete</td>
<td>Spinal motion</td>
<td>Kinematic</td>
<td></td>
</tr>
<tr>
<td>Yettram &amp; Jackman (1982)</td>
<td>2D</td>
<td>Discrete</td>
<td>Postural loading</td>
<td>Static, linear optimisation</td>
<td></td>
</tr>
<tr>
<td>Meakin et al. (1996)</td>
<td>2D</td>
<td>Continuous</td>
<td>Stability</td>
<td>Static and dynamic</td>
<td>Sagittal curvature</td>
</tr>
<tr>
<td>Scholten et al. (1988)</td>
<td>3D</td>
<td>Continuous</td>
<td>Stability</td>
<td>Non linear</td>
<td>Sagittal curvature</td>
</tr>
<tr>
<td>Shirazi Adl &amp; Parianpour (1996)</td>
<td>3D</td>
<td>Discrete</td>
<td>Stability</td>
<td>Non linear</td>
<td>Pelvic rotation, flexion moments</td>
</tr>
</tbody>
</table>

117
<table>
<thead>
<tr>
<th>Author</th>
<th>Focus of Study</th>
<th>Joint level</th>
<th>Muscle force estimation</th>
<th>Estimate of applied load</th>
<th>Additional features</th>
<th>2D/3D</th>
<th>Static/Semi-dynamic</th>
<th>Sagittal symmetric list</th>
<th>Transverse symmetric list</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carr &amp; Hering (1979)</td>
<td>2D Static</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Static</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Chaffin (1989)</td>
<td>2D Static</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Static</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Wood &amp; Lavy (1980)</td>
<td>2D Semi-dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Semi-dynamic</td>
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<td>transverse symmetric list</td>
</tr>
<tr>
<td>Prieto et al. (1984)</td>
<td>2D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Dynamic</td>
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</tr>
<tr>
<td>Lederman et al. (1989)</td>
<td>2D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
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<td>transverse symmetric list</td>
</tr>
<tr>
<td>Bush-Joseph et al. (1988)</td>
<td>2D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>McGill &amp; Norman (1987)</td>
<td>2D Semi-dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
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<td>Semi-dynamic</td>
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<td>transverse symmetric list</td>
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<tr>
<td>McGill &amp; Norman (1985)</td>
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<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Static</td>
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<tr>
<td>Schulze &amp; Anderson (1981)</td>
<td>3D Static</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Static</td>
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<td>transverse symmetric list</td>
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<tr>
<td>Marras &amp; Reilly (1988)</td>
<td>2D Static</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
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<td>Bone &amp; Chaffin (1989)</td>
<td>2D Semi-dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
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<tr>
<td>Han et al. (1991)</td>
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<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Marras &amp; Sommerich (1995)</td>
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<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>2D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Marsa &amp; Manas (1995)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Collet (1993)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Ogata et al. (1996)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Ogata &amp; Ogata (1999)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Chaffin et al. (1992)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Chaffin et al. (1998)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
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<tr>
<td>Bongio et al. (1999)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
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<tr>
<td>Bongio et al. (1995)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Stokes &amp; Gardner (1995)</td>
<td>3D Semi-dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Semi-dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Bergmann (1999)</td>
<td>3D Static</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Static</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Chaffin &amp; Ogata (1992)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Shahab &amp; Parpaul (1993)</td>
<td>3D Dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Areyan (1999)</td>
<td>3D Semi-dynamic</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Semi-dynamic</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
<tr>
<td>Laverne et al. (1992)</td>
<td>3D Static</td>
<td>L/2/5</td>
<td>Muscle force estimation</td>
<td>Analysis of motion film</td>
<td>Methods of lifting</td>
<td>3D</td>
<td>Static</td>
<td>sagittal symmetric list</td>
<td>transverse symmetric list</td>
</tr>
</tbody>
</table>
Table 2-5: Summary of the main features of the unit spine models discussed in this chapter.

<table>
<thead>
<tr>
<th>Authors</th>
<th>Type of unit</th>
<th>Geometric Representation</th>
<th>Focus of study</th>
<th>Vertebreal bone</th>
<th>Facets</th>
<th>Ligaments</th>
<th>Discs</th>
<th>Loading</th>
<th>Features</th>
</tr>
</thead>
<tbody>
<tr>
<td>Teo et al. (1994)</td>
<td>C2 vertebra</td>
<td>Axysymmetric</td>
<td>Effects of head Impact on C2</td>
<td>Cortical bone only</td>
<td>Spring elements</td>
<td>Spring elements (transverse ligaments)</td>
<td>-</td>
<td>45, 0, -45 force direction in sagittal plane</td>
<td>Odontoid process</td>
</tr>
<tr>
<td>Simon et al. (1985)</td>
<td>VB-disc-VB</td>
<td>Axysymmetric</td>
<td>Permeability of disc</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Poroelastic</td>
<td>Axial compression</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Suwito et al. (1992)</td>
<td>VB-disc-VB</td>
<td>Axysymmetric</td>
<td>Parametric study of material and geometric properties</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>No ground substance</td>
<td>-</td>
<td>Vertebral body curvature included</td>
<td>-</td>
</tr>
<tr>
<td>Oxland &amp; Onat (1993)</td>
<td>Superior half T12 vertebra</td>
<td>Axysymmetric</td>
<td>Thoracolumbar Burst fracture mechanism</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Pressure on VB</td>
<td>Axial compression</td>
<td>Disc degeneration</td>
<td>-</td>
</tr>
<tr>
<td>Natali (1991)</td>
<td>VB-disc</td>
<td>Axysymmetric</td>
<td>Disc Dehydration and Degeneration</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Hyperelastic</td>
<td>Axial compression</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Balasubramanian et al. (1979)</td>
<td>Vertebral</td>
<td>Bilaterally symmetric</td>
<td>Laminectomy</td>
<td>Cortical, cancellous and bony end plate</td>
<td>Hexahedral elements</td>
<td>-</td>
<td>Pressure on superior end plate</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Bozic et al. (1994)</td>
<td>C4 vertebra</td>
<td>3D</td>
<td>Burst fracture of cervical vertebra</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Stiffness as springs</td>
<td>Uniform compressive axial displacement</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Faulkner et al. (1991)</td>
<td>VB</td>
<td>3D</td>
<td>osteoporosis</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Axial compression</td>
<td>Yield stress prediction</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Kaara et al. (1991)</td>
<td>VB-disc-VB</td>
<td>3D and axysymmetric</td>
<td>Vibration</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Ground substance</td>
<td>Step and harmonic axial loading</td>
<td>Compressive preload, disc degeneration</td>
<td>-</td>
</tr>
<tr>
<td>Lu et al. (1996)</td>
<td>Motion segment</td>
<td>3D</td>
<td>Disc prolapse</td>
<td>Cortical, cancellous and bony end plate</td>
<td>2D cable elements</td>
<td>Ground substance</td>
<td>Combined compression, flexion and axial rotation, Viscelastic discs and ligaments</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Mizrahi et al. (1993)</td>
<td>VB</td>
<td>Bilaterally symmetric</td>
<td>Osteoporosis</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Axial compression and forward flexion</td>
<td>Material and geometric parameters</td>
<td>Material and geometric parameters</td>
<td>-</td>
</tr>
<tr>
<td>Mizrahi et al. (1992)</td>
<td>VB</td>
<td>Bilaterally symmetric</td>
<td>Metastatic lesions</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Axial compression and forward flexion</td>
<td>Material and geometric parameters</td>
<td>Material and geometric parameters</td>
<td>-</td>
</tr>
<tr>
<td>Shirazi Adl (1984)</td>
<td>VB-disc</td>
<td>Sagittal and mihdorizonal plane symmetry</td>
<td>Disc representation</td>
<td>Cortical, cancellous and bony end plate</td>
<td>-</td>
<td>Ground substance</td>
<td>Axial compression</td>
<td>First to include ground substance</td>
<td>-</td>
</tr>
<tr>
<td>Shirazi Adl et al. (1986)</td>
<td>Motion segment</td>
<td>Bilaterally symmetric</td>
<td>Disc and facet representation</td>
<td>Cortical, cancellous and bony end plate</td>
<td>Non linear contact</td>
<td>Axial elements</td>
<td>Ground substance</td>
<td>Flexion and extension</td>
<td>-</td>
</tr>
<tr>
<td>Shirazi Adl &amp; Droolin (1987)</td>
<td>Motion segment</td>
<td>Bilaterally symmetric</td>
<td>Disc and facet representation</td>
<td>Cortical, cancellous and bony end plate</td>
<td>Non linear contact</td>
<td>Axial elements</td>
<td>Ground substance</td>
<td>Single and combined sagittal loading</td>
<td>-</td>
</tr>
<tr>
<td>Shirazi Adl (1991)</td>
<td>Motion segment</td>
<td>Bilaterally symmetric</td>
<td>Disc and facet representation</td>
<td>Cortical, cancellous and bony end plate</td>
<td>Non linear contact</td>
<td>Axial elements</td>
<td>Ground substance</td>
<td>Combined compression, flexion and axial rotation</td>
<td>Compressive preload</td>
</tr>
<tr>
<td>Shirazi Adl (1989b)</td>
<td>Motion segment</td>
<td>Bilaterally symmetric</td>
<td>Disc prolapse</td>
<td>Cortical, cancellous and bony end plate</td>
<td>Non linear contact</td>
<td>Axial elements</td>
<td>Ground substance</td>
<td>Combined loading: symmetric and non symmetric lifting</td>
<td>Progression of disc rupture</td>
</tr>
</tbody>
</table>
The previous classification scheme by Goel and Gilbertson (1995) considered finite element models as either simple or complex. For example the model of Stokes and Gardner-Morse (1991, page 56) of the thoraco-lumbar spine in which each motion segment was represented by a single finite element was classified as simple in comparison with very detailed models of an individual disc and vertebrae. However, this does not reflect how this finite element model compares with other models of the thoraco-lumbar region. When considered using the classification scheme presented in this paper, the model appears as a highly advanced model for that region and for its era. The way models represent structural components and the way they investigate an application common to a given region of the spine can thus be appreciated using this classification scheme. For each region, the features of the models and the significant contribution which they have made to spinal research is highlighted.

The large mass of the head in the cervical region, acting anterior to the vertebral centres, has considerably larger inertial properties than the cervical vertebrae which support it. Sudden applications of a load, typical of impact situations such as vehicle collision creates large bending moments in the spine and increases the risk of injury. Models of this region have thus mainly been concerned with this problem.

The thoraco-lumbar region of the spine is mainly responsible for bearing the weight of the trunk and any loads carried. Several models of this region have investigated the problem of scoliosis which impedes this load bearing function. Some of these have included representation of the complex geometric and material properties of the ribcage.
The numerous models of the lumbar region have been developed mainly to investigate the causes of low back pain. This has included prediction of the loads acting on the spine for a variety of lifting strategies, and investigation of microscopic symptoms such as disc degeneration and osteoporosis using finite element techniques. Some finite element models have also been used to investigate the effectiveness of surgical correction devices at several levels of the spine.

While there exists a general trend for the level of detail of the model to increase with a reduction in the region of the spine modelled, the further development of computational technology may alter this. In particular, finite element studies have begun to be developed for the whole lumbar spine so that the effects of injury, degeneration and instrumentation may be analysed at several levels, while models including low back muscles have investigated the effects of multi-segmental muscles. The advances in modelling techniques for the lumbar region, and indeed the number of models developed are considerable.

These modelling techniques are now being applied to other regions of the spine. Finite element models of the cervical spine have been developed to investigate the stress distributions associated with impact loading. These studies are however limited by the material properties available. Also, lumbar modelling techniques for the prediction of muscle forces are being applied to the cervical region. For example the model of Schultz and Andersson (1981, page 68) which predicted muscle forces acting at the L3 level was adapted by Moroney et al. (1988) for the cervical region. Again however, the array of data for the muscles of the lumbar region is not yet available for the cervical (or thoracic) regions.
While modelling techniques may continue to improve, their validity depends on their agreement with known behaviour, and therefore they cannot replace experimental techniques. In particular, models developed for the lumbar regions including muscle activity have relied heavily upon experimental input. These have included estimates of body positions and movement using radiographic displacements, light emitting diode markers (LED), accelerometers and more importantly estimates of muscle forces using CT scans and magnetic resonance images, and EMG data.

The numerous models developed for the lumbar regions of the spine emphasise the concern in spinal research for the existing problem of low back pain. However, discrepancy exists between the focus of investigation and the degree of anatomic detail. Consequently, it is difficult to compare the findings of these studies, and the causal mechanisms of low back pain have not yet been agreed upon.

Simple lever models which predict the loads at a single inter vertebral joint are appealing for the investigation of tasks in the workplace due to their simplicity. Indeed guidelines concerning load limits and lifting techniques in industrial tasks have been based upon these model predictions (for example 'NIOSH guidelines' (1981) in the US and the 'UK Guidelines on the European Directive on Manual Handling' (HSE, 1992) in the UK). However, these models assume the greatest loads are associated with the inter vertebral level of investigation. Ascending the spine, smaller forces due to body weight are applied, but are matched by smaller vertebral body cross section and smaller muscle cross section. The balance between muscle and body weight forces at each and every level of the spine is equally important.
In addition, the level of anatomic detail of a model has been shown to considerably affect the accuracy of its predictions (McGill and Norman, 1987, page 67). This is reflected in the predictions of simple lever models which predict inter vertebral joint reactions to exceed the known strength of the vertebrae.

Multi-segmental models (page 81) generally include a greater number of muscle forces, and include a greater degree of anatomic detail. This however increases the indeterminacy of the spine. Prediction of the muscle forces using optimisation allows a vast number of forces to be included, but is restricted by the objective function which does not necessarily represent the biological recruitment strategy for muscles. In contrast, EMG based models are limited practically by the number of muscles they can monitor, but provide muscle recruitment patterns unique to the individual. A combined optimisation and EMG approach (Cholewicki and McGill, 1995, page 76) has been reported to be a more favourable alternative.

Multi-segmental muscle models also allow the stability of the spine to be determined, and have included a role for the deeper segmental muscles of the spine which was not considered by previous single level models. However, studies of EMG patterns and cross sectional areas have not yet explored these muscles. These models are therefore limited by data regarding the force generating capacity of the muscles.

Multi-segmental models have also allowed the effects of curvature to be considered. Finite element models have revealed lumbar lordosis to result in coupled motion of the spine in response to a pure applied force or moment. Disc and facet degeneration at one level have also been considered for their effects at other levels, and were found to be significant. The outcome of these studies emphasise the interactive nature of the spinal system, and how a change in a single component is likely to affect the system as a whole.
A greater role for the curvature of the lumbar and thoraco-lumbar spine was proposed when representing the spine as an arch (Aspden, 1987, page 89). This model requires the prediction of a thrustline which provides an indication of the state of equilibrium and stability of the system. The model predictions implied the curvature of the spine reduces the muscle forces required for stability. However the degree of anatomic detail in this model does not match that of other lever-based multi-segmental models.

Thus, while models are powerful tools for spinal research, the findings of this review indicate models vary significantly in their applications, the components modelled and the level of detail included. Based upon this review, the development of a new mathematical model is now considered.
2.6. THE PURPOSE OF A NEW MATHEMATICAL MODEL

The prevalence of the low back pain, and the costs associated with it clearly show low back pain to be an area of major concern to industry, and to the population of many Westernised societies. Investigations to identify the underlying causes of low back pain, and to develop possible prevention strategies would therefore be beneficial both to society and to the individuals at risk. The direction of this work was therefore to continue the existing research into the underlying causes of low back pain.

Mathematical modelling techniques clearly show considerable advantages over experimental techniques for investigating low back pain. In particular, mathematical models offer the flexibility to investigate hypothetical situations and to try out new ideas for the causes of low back pain. The development of a mathematical model which exploits this feature was considered an attractive medium for these investigations.

Although low back pain has many possible underlying causes, mechanical factors appear to play a significant role; particularly jobs such as manual materials handling or nursing which involve lifting large loads in constrained work places which are associated with a high incidence of low back pain. Mathematical models are considered particularly relevant for determining the loads involved.
In particular, the magnitude of the load and the way it is lifted are the two main variables which influence spinal loading. The applied load includes the external load carried, and the forces due to body weight. The way these are transmitted through the spine, and the consequent loads on the spine at each vertebral level depends upon lifting technique. In turn, lifting technique is dependent upon posture, which is ultimately dependent upon spinal configuration. The configuration of the spine during lifting is therefore very important.

The configuration of the spine is determined by the position and orientation of each vertebrae. These can be considered as a continuous series whereby a change in direction of one vertebra has the potential to affect all others. Consideration of the whole spine is therefore essential.

The control of the vertebral positions is ensured by forces generated by the ligaments and muscles along the entire length of the spine. These forces also affect the consequent loads at each inter vertebral joint. An imbalance in force at any one level may cause a change in position of the vertebrae at one level, and consequently at several other levels, thereby affecting the whole system. Consideration of the configuration of the spine, and how its curved path influences loading requires consideration of the forces acting at every inter vertebral level.

The literature review reveals numerous models which investigate lifting techniques. However, they only consider equilibrium of forces and moments about a single inter vertebral joint. This is usually one of the lower lumbar inter vertebral joints on the basis that loads are considered to be greatest at this level. The role of the configuration of the spine has therefore largely been ignored.
The focus of this work was therefore to develop a mathematical model of the spine which would allow the relationship between spinal configuration and the consequent loads to be investigated.

In particular, Aspden has proposed the curvature of the spine may have a role in load bearing when the spine is represented as arch structure. This involves both muscle forces and spinal curvature acting together to withstand the loads applied. Indeed the arch model may be critical towards gaining a greater understanding of the role of the curvature of the spine in load bearing, and the effect it may have upon the incidence of low back pain (Grilli and Acar, 1997).

The predictions of Aspden’s model show a significant reduction in the predicted loads at the L5/S1 inter vertebral joint when representing the curvature of the spine as an arch. However the existing arch model includes only simple loading patterns for body weight and muscle forces, and only partial representation of the spine. The degree of anatomic detail in this model therefore does not match that of lever-based multi-segmental models. Indeed for the lever models the level of anatomic detail of a model has been shown to considerably affect the accuracy of its predictions (McGill and Norman, 1987). A similar effect is expected for the arch model. Also, the role of the muscles in ensuring equilibrium at each level of the spine cannot be considered in this simplified form.

The arch model also provides an indication of stability for the spine, determining whether the path of the resultant forces is transmitted between adjacent vertebrae. This may provide insight into the roles of individual muscles which the lever model does not. The role of the curvature of the spine in load bearing, and its relationship between muscle forces was therefore considered a key feature of the model to be developed.
Modelling the spine in this way allows a variety of postures and loading conditions to be investigated. The outcome can be used to identify postures which minimise inter vertebral compression and muscle forces for a given activity, thereby predicting 'safe' techniques with which to work. Also the design of chairs which encourage the load bearing role of the spinal curve may be vital in reducing the incidence of low back pain reported for sedentary office activities and long distance driving. The model may also be used to investigate the changes in curvature and increased loading on the spine during pregnancy. In addition to identifying correct postures, exercises can be recommended to help strengthen the muscle forces responsible for control or maintenance of the spinal curvature.

The purpose of this study was therefore:

1) To continue the existing necessary research in determining the underlying causes of low back pain and back pain in general.

2) To exploit the flexibility with mathematical modelling techniques offer, in investigating new hypotheses.

3) To develop further investigation into the role of the curved configuration of the spine using greater anatomic detail and to indicate its importance as a future element of research.

4) To investigate the roles of the individual muscles when representing the spine as an arch and to provide an indication of their contribution towards both stability and spinal loading.
CHAPTER THREE:

3. DEVELOPMENT OF A NEW MATHEMATICAL MODEL OF THE HUMAN SPINE

The main requirement of the developed model was that it would allow the relationship between spinal curvature, muscle and ligament forces and the consequent loads to be investigated. The underlying concept of the arch model was considered to be a suitable basis for these investigations. However, the anatomic detail of the existing arch model required extensive developments. In order to identify these developments, the existing model and its underlying theory were considered.

The three main developments were to include the curvature of the whole spine, forces due to body weight acting at each vertebral level and forces generated by the individual muscle groups. The relations between curvature and the applied forces on the spine are described using algorithms in a computer based modelling application (ADAMS). The output of this model provides information regarding the joint reaction forces at each level and the thrustline co-ordinates. These can provide an indication of the loading and stability for the spine in a given posture.

The input data to this model requires descriptions of body weight forces, muscle forces and the curvature of the spine in various postures. This is derived based upon the available data in the literature.
3.1 EQUILIBRIUM AND STABILITY OF AN ARCH STRUCTURE

For a structure to be represented as an arch, its components must be compressed together sufficiently in order for the arch to retain its shape and prevent collapse when subjected to applied loads. Mechanically, the arch must be in equilibrium and stable (Heyman, 1982).

3.1.1 EQUILIBRIUM

The reference line of an arch can be defined as that joining its two ends. For an arch to be able to withstand external loads, forces must be generated at either end of the arch. The component of force parallel to the reference line is known as the compressive thrust (H) and helps provide strength and stiffness to the arch. Opposing directions of this thrust at either end ensures equilibrium in this direction, while the perpendicular force component (R) satisfies equilibrium with those of the external loads acting on its surface. The resultant forces acting at each point of loading on the arch collectively follow a path known as the thrustline. The existence of this thrustline represents equilibrium.

The example in Figure 3-1 shows three external forces (W1, W2 and W3) acting on the convex surface of a symmetric horizontal arch.
For the arch to be in equilibrium, two conditions are necessary:

i) the sum of all the forces acting vertically and horizontally is zero;

ii) the sum of the moments generated by these forces about any point within the arch must be zero.

To determine the components of the reaction forces generated at the ends of the arch (A and B), moments are considered about these points. The line of action of the parallel component (H1 and H2) passes through both A and B and therefore generates zero moment about these points. Only the perpendicular reaction force components (R1 and R2) can therefore be determined in this way.
For a prescribed set of external loading conditions, the perpendicular reaction components are fixed in order to satisfy equilibrium. However, as long as the values of H1 and H2 are equal and opposite (thereby satisfying equilibrium), the range of values of H1 and H2 which satisfy the equilibrium condition is infinite. These values are however limited by requirements for stability. For a masonry arch it is ensured that the end supports are strong enough to generate the necessary reaction forces to ensure equilibrium and stability for a pre-defined range of applied loads.

3.1.2 STABILITY

Stability of the arch is achieved when the resulting forces acting on the arch can be transmitted between adjacent elements along its entire length. For this to occur, the thrustline must lie within the cross section of the arch. The range of values for H1 and H2 for stability is therefore finite and is restricted to small limits, depending upon the applied loads and the curvature of the arch.

The funicular polygon is a graphical approach for determining the thrustline of an arch structure for given loading conditions. Mathematically, this method requires the force components parallel to and perpendicular to the reference line to be determined and the direction (θ) relative to the reference line of the resultant vector (w) calculated (Figure 3-2).
The angle each resultant vector makes relative to the reference line effectively determines the slope of the thrustline for each section defined between two adjacent points of loading. Consequently, the overall path of the thrustline depends upon the direction of the resultant force acting in each section of the arch (Figure 3-3).

Thus, the existence of a thrustline indicates equilibrium of the structure and the path of this thrustline within the cross section of the arch along the entire length of the spine indicates stability. Equilibrium is therefore a necessary but not sufficient condition for stability.
Several thrustlines may exist which satisfy the stability condition, depending upon the proportions of the perpendicular and parallel components. Transmission of a force outside the limits of the arch may result in the formation of a hinge and ultimately cause collapse. Therefore according to the safe theory of arches, the thrustline which passes closest to the centre of the arch is the optimal and forces which pass within the middle third of the arch are acceptable (see Heyman, 1982).

For loads acting on the convex surface of the arch in a direction perpendicular to the reference line, the reaction force component at the end supports is large and the thrustline is more inclined. In contrast, for loads acting mainly parallel to the arch reference line, the thrustline gradient is less. Designers of masonry arches calculate the cross section of the arch and the required curvature necessary to withstand loads due to vehicular or human motion within a known range. For larger loads, a greater curvature is needed to coincide with the greater inclination of the thrustline.
3.2 ASPDEN'S APPLICATION OF ARCH THEORY TO THE SPINE

Aspden (1987, 1988, 1989) has applied the theory of arch structures to the spine in various postures and proposed that the curvature can assist in load bearing.

3.2.1 CONDITIONS OF ARCH THEORY

In the flexed posture, the lordotic curve of the lumbar spine may flatten and reverse to form a single anterior concave curve with the thoracic spine. This configuration represents the curvature of an arch in which the reference line is near horizontal. Forces due to body weight which act vertically downwards with gravity therefore act approximately perpendicular to its convex surface.

Aspden therefore proposed that the curve of the spine and the direction of the applied loads were analogous to the loading conditions of a masonry arch. The compressive force needed to compress the elements of the spine together and to maintain the configuration under the applied loads could be therefore calculated using the theory of masonry arches. This was proposed to be provided by the reaction force at the sacrum acting as an end support and by a collective force representing the weight of the head and muscle and ligament forces acting at the other end in a direction mainly parallel to the spine. For this structure, the muscle and ligament forces act intrinsically to the spine.
For stability, Aspden proposed the thrustline to lie within the middle third of the spinal arch corresponding approximately to the region of the vertebral bodies. The magnitude of the muscle and ligament forces could be varied in order to accommodate an increase in applied external loads and to ensure stability of the spine. Aspden (1989) demonstrated this by considering changes in posture.

In order to represent the spine as an arch, certain requirements had to be satisfied:

i) Sliding movement between the elements of the arch must be negligible. Aspden therefore assumed that the shear forces between the vertebrae can be withstood by the intervertebral discs.

ii) The safe theorem assumes that the arch model is able to withstand compressive forces only. Tensile properties of the disc are therefore not considered.

iii) The arch has infinite compressive strength. Aspden reported the model predictions for the vertebral compression were less than their known failure strengths and also that failure of one vertebra does not result in collapse of the entire spine.
3.2.2 POSTURES INVESTIGATED

In the erect posture the lordotic curve of the lumbar spine was considered. Components of force due to body weight run almost parallel to the arch reference line and therefore contribute to the compressive thrust. The force required by the muscles in this situation is therefore reduced and a relatively flat thrustline path is obtained. By comparison, in the flexed posture the forces due to body weight and the lifted load act almost perpendicular to the convex surface of the thoraco-lumbar spine. The compressive force required to be provided by the muscle forces for stability is therefore increased. Due to this demand for greater muscle activity, the compressive force on the intervertebral joint is higher.

Aspden performed a plastic analysis for the spine in the postures described above. Although several combinations of muscle forces can be found which satisfy stability of the spine as an arch, Aspden proposed the generation of at least one thrustline which corresponds to the path of the vertebral bodies and indicates that the spine is stable in that position. The unknown forces due to body weight, muscles and ligaments were therefore lumped together as a single force at the upper end of the arch. In this way an assessment of the compressive force required to be generated by body weight, muscle and ligament forces and the resulting loads on the intervertebral joints could be calculated for a given posture. By comparing the predicted reaction forces at the L5/S1 intervertebral joint with those predicted by a lever model of the same loading conditions, an assessment of the load bearing role of the curvature could be made.
Aspden (1988) reported the model is particularly useful in assessing the demands on muscle forces and the intervertebral joints in postures which are determined by confined work spaces or loads which require particular lifting strategies. In particular, the model involving only three externally applied forces and a collective muscle force allows a very simple analysis of tasks. Further application of this model to a variety of postures and loading conditions could be used to identify optimal working conditions.

However, for the lever model a simple anatomic representation resulted in an over-estimation of the predictions of the loads at the L5/S1 intervertebral joint. McGill and Norman (1987) in their model showed a more detailed representation of the spinal muscles reduced the joint reaction predictions. The predictions of the existing arch model can therefore not be applied to in vivo situations without investigating the effects of more detailed force patterns.
3.3 FEATURES OF THE NEW MODEL

The new mathematical model developed is based upon the underlying concept of arch theory but includes a greater degree of anatomic detail. These features are also discussed in Grilli and Acar (1997a).

3.3.1 WHOLE SPINE

Previous investigations involved representation of the lumbar spine in the erect posture and the thoraco-lumbar spine in the flexed posture, each as a single arch. However stability of one particular region of the spine does not imply stability in other regions. Realistically therefore, stability of the whole spine must be considered.

Simulation of a lifting activity requires the range of configurations from a fully flexed position to an upright posture at the end of the lift be considered. However, representation of the spine as a whole involves three distinct alternating curves in the erect posture (Figure 4). When the spine flexes, the posterior convex curve of the lumbar region may flatten or reverse. Application of arch theory to the whole spine therefore requires adaptability to any configuration with any associated curves. This requires consideration of the three curvatures of the spine in the erect posture, and how these curves change with flexion.
3.3.1.1 Curvature

In the erect posture, the cervical and lumbar curves are anteriorly convex and the thoracic curve is posteriorly convex. Although the vertebrae of the spine possess certain common geometric features, variation in each region reflects the range of movement possible at each level. The geometric properties of the vertebrae in the transitional regions between the curves also vary. However, no other geometric distinctions marking the ends of each curve can be detected.

In flexion, the distinction between these curves is diminished as the lumbar and cervical curves flatten and possibly reverse to join the flexed thoracic curve. The degree of flexion between each vertebrae is not uniform, and individual variation accounts for the infinite range of postures which can be configured.
The spine is not therefore three separate independent curves which can each be represented as an arch structure. Rather the spinal curve is a single continuous entity where a change in configuration at one level can result in a change in configuration at all other levels. The term spinal arch used in this thesis refers to this variable curve.

The compressive thrust generated for the whole structure is due to a combination of body weight, muscle and ligament forces. Adjustments in the thrustline path so that it conforms to the alternating curvature of the spine in the erect posture is possible by contraction of the muscle and ligament forces distributed along the entire length. Further, these muscle elements are active and may change their contractile state to suit the direction and the level of compressive force required. Unlike the masonry arch whose shape has been calculated and fixed in order to ensure stability, the spine is a dynamic structure, which may adjust muscle recruitment patterns or curvature to ensure equilibrium and stability for a range of loads and postures.

3.3.1.2 End Supports

Unlike masonry arches which have strong abutments securing their connection with the ground, the spine has no such supports. Also, whereas masonry arches lie approximately horizontal, the spine often functions in a more upright manner, and is in the erect posture, almost vertical.

The sequential arrangement of the vertebrae is such that forces applied at the upper end have a cumulative effect at the lower end. As with Aspden's model, the superior surface of the sacrum is considered to represent the lower end of the spinal arch. The reaction force generated at this end ensures equilibrium of forces, and generates the required thrust at this end (Figure 3-5).
The upper end of the spine is considered to be the superior surface of the atlas vertebra. For equilibrium of the head about the spine, muscle and ligament forces are required. The compressive thrust provided at the upper end of the spine is therefore due to the components of force as a result of head weight and neck muscle and ligament forces (Figure 3-5). However, due to the distribution of force at this end of the spine, the compressive thrust provided is considerably smaller than at the lower end of the spine. The active control by the muscles along the entire length of the spine ensures that the necessary compressive thrust is generated to hold the elements together, and to satisfy conditions of equilibrium and stability for an arch.

Figure 3-5: Possible means by which the compressive thrust at the ends of the arch is applied.
3.3.2 DISTRIBUTED FORCES DUE TO BODY WEIGHT

For consideration of the loads on the spine in various daily postures, the weight of the upper body must be considered. Aspden's investigations (1989) were based upon the same loading conditions as those for the lever model of Morris et al. (1961), so that a direct comparison between the two models could be made. This involved representing the force due to the weight of the head and upper body applied at T2, and the force due to the weight of the trunk applied at T12. These forces were assumed to act at the vertebral centres.

However, the detailed curvature of the spine ideally requires a loading pattern which will control the path of the thrustline at each level of the spine (Grilli and Acar, 1997b). An increase in the number of loading points results in a greater number of links in the thrustline, and a smoother thrustline path. Also the forces due to body weight do not act directly upon the centres of the vertebral bodies, but act eccentrically to each vertebra. This must be accounted for when deriving the distributed loading pattern.

3.3.3 DISTRIBUTED FORCES DUE TO MUSCLES AND LIGAMENTS.

In addition to the simplified representation of body weight, Aspden's investigations (1989) involved a collective force to represent the action of muscle and ligament forces (and the weight of the head). However, realistic muscle and ligament forces are generated along the entire length of the spine.
The points of attachment of the muscle groups at certain levels allows them to perform the necessary local or major adjustments in curvature so that the resulting thrustline corresponds to the vertebral configuration. Distribution of the muscle forces along the spine allows this localised control, and also allows individual control of the loads at each intervertebral level (Grilli and Acar, 1997b).

The muscles and ligaments also vary in their point of attachment on the vertebrae and discs, the magnitude of force they can generate, the line of action of these forces and the moment arm. These factors cannot be considered when a single collective force is applied at the upper end.

The spinal system is highly indeterminate and the precise muscle recruitment strategy for a given posture is unknown and varies with each individual. When representing the muscles collectively, this problem is not considered. Instead Aspden (1988) proposed that the muscle force which could result in at least one thrustline within the cross section of the spine at all levels indicated stability for a given posture.

The aims of this study were to determine if stability of the spine could be satisfied by the application of forces generated by individual muscle groups, accounting for the variation in direction, points of application and magnitude. A resulting thrustline within the cross section of the spine at all levels for a particular combination of muscles may not represent the precise in vivo muscle recruitment strategy but implies the muscle forces are sufficient to satisfy conditions of stability for the given posture.

The individual roles of the muscles in adjusting spinal curvature or the magnitude of the compressive force, and the consequent effect on the loading at the intervertebral joints can therefore be investigated using a distributed loading pattern.
3.3.4 VARIATION IN POSTURE

A change in spinal configuration alters the direction of body weight forces relative to the reference line and affects the amount of muscle and ligament force required for stability. The active components of the muscles allow them to respond by changing their contractile state to suit the direction and level of the compressive force required.

A change in spinal configuration can also alter the relative locations of the points of muscle attachment on the vertebrae and the consequent line of action. This can be a direct change in direction of muscle force relative to the reference line, or a deviation of the line of action of the muscle so that it follows the curvature of the spine. The moment arm of the muscle force can also be affected.

In this study, four postures representing various stages of a lifting activity are considered. The effects of changes in curvature on the amount of muscle force required for stability, and the force generating capacity of the muscles and ligaments can be determined. Consequent changes in the thrustline can also be determined.
3.3.5 INTERVERTEBRAL JOINT REACTIONS

Forces due to body weight increase descending the spine. Typically the lumbar vertebrae demonstrate the largest cross sectional area while the cervical vertebrae are considerably smaller, reflecting their load bearing requirements. However, instability may occur at any level of the spine. With a distributed muscle force pattern, an increase in force of a particular muscle group to control the thrustline at a certain level may result in an increase in force at a particular level. This force may be disproportional to the strength of the vertebra and disc at that level. Determination of the loads at each intervertebral joint is therefore necessary.

The three assumptions made by Aspden (p136) so that the spine satisfied the conditions of the arch theory are to be continued in the developed model. Assumptions that the shear resistance of the intervertebral disc is sufficient to prevent sliding failure, and that the vertebral compressive strength is sufficient to withstand the applied loads are to be determined by considering the level of loading predicted by the model.

3.3.6 TWO DIMENSIONAL REPRESENTATION

Investigations performed using the arch model with collective loading assumed symmetrical postures and forces about the sagittal plane. In order that the effects of increasing the number of muscle and body weight forces could be evaluated, this two dimensional representation was maintained.
Due to the number of forces involved, it was considered necessary to reduce any unnecessary complexity of the model. According to the safe theorem resultant compressive forces are required to be transmitted through the vertebral bodies. The posterior elements of the spine were therefore not represented geometrically and a simple two dimensional block representation of the vertebral bodies was used. A previous parameterised geometric model of the human spine (Stepney et al. 1996) could later be used to provide a good geometric representation when the mathematical model had been developed.

3.3.7 MODEL APPLICATIONS

Previous studies (e.g. McGill and Norman, 1985) have shown models which include acceleration and inertia forces associated with dynamic lifting predict a greater compressive component for the intervertebral joint reaction. Ultimately, extension of the existing arch model to consider a dynamic situation is therefore required. However, in order to appreciate the effects of greater anatomic detail (through distributed body weight and muscle force patterns for the whole spine), it is necessary to restrict the number of variables in the model so that the effects of the anatomic detail alone can be considered. By assuming that the lifting activity is performed slowly, acceleration and inertia can be omitted. In this study, four postures representing various stages of the lifting activity are therefore investigated under static loading conditions.
3.4 AIMS AND OBJECTIVES

In section 2.6, the proposal for a new model which investigated the role of the curvature of the spine was outlined. In particular two main aims were identified:

- To pursue investigations into the role of the curved configuration of the spine using greater anatomic detail, and to indicate its importance as a future element of research.
- To relate model predictions of stability and loading of the spine for a given posture to possible causes of low back pain.

Consideration of the previous arch model (Aspden, 1987) has indicated that greater anatomic detail must be considered and that this should include:

i) curvature of the whole spine;
ii) forces due to body weight acting at each vertebral level;
iii) forces generated by the individual muscle groups.

Through comparison of model predictions for loading and stability with those using collective loading conditions, the effects of this anatomic detail can be appreciated. This includes:

1) Comparison of stability of the spine in the postures investigated.
2) Comparison of the amount of muscle force required to ensure stability.
3) Comparison of the predicted joint reaction forces due to muscle activity in the stable postures.

Comparison of the model predictions with the strengths of the vertebrae and individual muscle groups can also be made to indicate areas of tissue overload and possible causes of back pain.
Due to the complexity of the human spine, computer based modelling tools were considered essential for extensive investigation of the interactive role of the curvature and muscles of the whole spine. A number of possible benefits were identified:

i) The curvature of the whole spine requires definition of the position and orientation of twenty four constitutive moveable vertebrae. Changes in configuration require calculation of translation and rotation vectors which computer based modelling tools automatically perform.

ii) A vast array of muscle and ligament forces act upon the curvature of the whole spine. Their line of action depends upon the configuration of the vertebrae to which they are attached and over which they span. Changes in configuration therefore require calculation of changes in muscle and ligament force vector components. This could be performed automatically by computer based modelling tools by coding the relation of these forces to vertebra position.

iii) The equilibrium and stability of an arch structure requires determination of the slope of the thrustline for the total number of forces acting. For \( n \) applied forces, there are \( n+1 \) sections of the predicted thrustline which represent different compression force vectors. Due to the number of forces acting upon the spine, a systematic way of defining the forces relative to the reference line of the arch and automating calculations for the resulting thrustline can be achieved with computer techniques.
iv) The total force acting on the spine at each level can be used to determine the corresponding intervertebral joint reaction. Due to the number of forces acting, this would require complex and tedious calculations for each of the 24 intervertebral joints. Calculations of this kind are also prone to human error. Computer based modelling tools would allow this procedure to be performed quickly and efficiently.

v) There is considerable biological variation between the material and geometric properties of the components of the human spine. Parametrization of these properties would allow the effects of variations to be determined. Using computer based modelling tools, these parametric tests could be performed systematically saving the modeller from tedious and repetitive calculations.

vi) The quantification of muscle forces and joint reactions provide invaluable information about the loads on the spine. Large quantities of data are however difficult to interpret. Graphical display within the application, or transfer of this information to other computer applications such as spreadsheets, allows this data to be represented in a more meaningful visual form.

vii) The development of a computer based model would provide a tool readily available for performing an unlimited number of investigations.
3.5.1 'ADAMS': A COMPUTER BASED MODELLING APPLICATION

The computer based modelling application ADAMS\(^1\) was identified as a suitable tool for developing the spinal model. The application was run on a Sun Sparc 20 work station, on which large data sets could be easily stored and processed. Network connections also allowed rapid transfer of data to a spreadsheet application available on PC's (Microsoft Excel). The compatibility which ADAMS provides with Unigraphics\(^2\) was considered suitable for including a good geometrical representation of the spinal components in a previous geometric model (Stepney et al., 1996). Further, the output of ADAMS may be written into a file structured for finite element applications for future stress analysis investigations.

ADAMS provides a means of visually designing and evaluating the system through its interactive graphical interface. Parts, markers, joints, forces and geometrical features are displayed as they are created, allowing the equivalent mathematical statements to be viewed. Hard copies may also be obtained for visual records. Through interaction with ADAMS VIEW, the user may at any time request information about an entity within the system, using the 'list info' option. The information required may be written to a separate information window which allows the user to read the information immediately or for further study and processing; the information may be written to a text file and can then be transferred to other applications such as spreadsheets if required.

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\(^1\) ADAMS (Automatic Dynamic Analysis of Mechanical Systems) produced by Mechanical Dynamics Incorporation (MDI)

\(^2\) Unigraphics produced by EDS (Electronic Data Systems Corp)
3.5.2 MATHEMATICAL REPRESENTATION OF THE SPINAL SYSTEM USING ADAMS

The purpose of the spinal model was to provide a quantitative investigation of the interactive role of the curvature and muscles of the whole spine. In response to a given configuration and set of applied forces, the model is required to predict the resulting thrustline and joint reactions.

This requires:
Definition of spinal configuration;
Definition of model boundary conditions;
Definition of forces;
Calculation and construction of thrustline;
Calculation of joint reactions;
Listing of joint reactions and thrustline co-ordinates.

The input to the model is therefore data regarding the spinal configuration and the bodyweight, muscle and ligament forces. Algorithms describing the relation between the spinal curvature and forces can then be used to calculate the thrustline and joint reactions. Output includes an indication of the stability of the spine through a comparison of the path of the thrustline relative to the positions of the vertebral bodies and the calculated loads on the joint at each level. A summary is given in Figure 3-6. Construction of the model depends upon a series of command lines written using the ADAMS command language. These commands are contained in several command files or macros which build the model each time they are called. An advantage is that storage of these text files occupy considerably less space than an alternative binary file which stores the current state of the model.
3.5.3 MODEL INPUT

Anatomic features regarding vertebral configuration and the applied body weight and muscle forces are represented in the model using variables (parameters). Separate macros were written to define data sets for each variable, so that the properties of a particular feature such as the vertebral length can be easily accessed and updated merely by changing the values in the data set. The data sets used in the investigations in this study are derived in section 3.6 (page 183).

In order to identify a particular variable with a particular vertebra, it was ensured the variables were assigned the same identification number as the corresponding part (vertebra) which they describe. The vertebrae from L5 to C1 were labelled sequentially from V1 to V24. Accordingly, all variables were suffixed by a number representing the vertebra number to which they refer (from 1 to 24). For example the variable massvar5 defined the mass of the L1 vertebra.
The model is constructed to predict the thrustline and joint reactions for body weight only by default. To determine the effects of additional muscle forces, the user enters the muscle group name, which calls upon the appropriate macro regarding the muscle force data. Several muscle groups can be entered.

In this way the input data regarding vertebra and disc height and length, vertebral orientation, the body weight acting at each vertebral level and the muscle forces, can be defined according to the requirements of the investigation.

3.5.3.1 Mathematical Representation of the Spinal Configuration

In this model, the centre of the superior surface of the sacrum was chosen to be the origin of the model and was positioned with global co-ordinates (0,0,0). Positioning of each vertebrae was defined relative to the vertebra below so that a change in configuration at lower levels would automatically change the configuration of the vertebrae above.

The origin of each vertebra was defined as the centre of the inferior endplate of the vertebral body. In the sagittal plane, the position of the vertebra above depended upon vertebral and disc height at the level below, illustrated in Figure 3-7.
Each vertebra was also assigned an angle of orientation. The sacrum, as the first part in the model, was oriented relative to the global y co-ordinate axis. In the upright postures this is approximately 40° (Chaffin, 1969). The orientation of the L5 vertebra was then defined relative to the sacrum and similarly for each vertebra, the orientation was defined relative to the axes of the vertebra directly below as illustrated in Figure 3-8.
The spinal configuration can thus be reconstructed from variables representing vertebra and disc height and relative angles of orientation with the position of each defined relative to the vertebra below.

Due to the origin of the sacrum at its superior surface and of all other vertebrae at the centre of their inferior surfaces, a separate description was needed to define the position of the L5 vertebra relative to the sacrum. This was determined by the height of the intervertebral disc in between.

For the remaining vertebrae, positioning of each vertebra was defined relative to the inferior surface of the vertebra below, according to the height of the inferior vertebra and the interconnecting disc. The x co-ordinate for each is zero.
The y co-ordinate of the origin of the superior vertebra relative to the co-
ordinate axes of the inferior vertebra can be defined in terms of variables
representing vertebra (vbheight) and disc height (dheight). For vertebra 'i'
(for i=2 to 24) the position is:

\[(\text{pos}_i = \text{vbheight}_i + \text{dheight}_i).\]

3.5.3.2 Geometrical Representation of the Vertebrae

The simple geometric representation for the vertebral bodies required the
creation of a simple block belonging to each part with dimensions
approximating the length (anterior-posterior dimension) and height of the
vertebral body in the sagittal plane. The location and orientation describes the
overall curvature of the spine. Parametrization of the vertebral body
dimensions, location and orientation requires a description for the creation of
only one block. By using conditional loops with different values for the
parameters at each iteration, the twenty four vertebral bodies of the spine can
each be represented constituting the spinal column.
3.5.3.3 Boundary Conditions for the Model

The applied forces in the spinal model include body weight, external loads lifted and muscle and ligament forces. Forces acting between the end limits of the arch directly affect the gradient of the thrustline following their point of application. However, forces acting at the upper boundary of the arch are also necessary to provide the compressive thrust at this end. Some of these forces also act indirectly by generating moments on the spinal structure system. It is therefore necessary to distinguish between forces which have a direct action upon the spinal arch and forces which lie beyond these boundaries but still have an effect upon the spinal system.

3.5.3.3.1 Arch Limits

The ends of the arch are defined to be the superior surface of the sacrum at the lower end and the superior surface of the C1 vertebra at the upper end. These can be represented by markers A and B respectively located at the centres of the two surfaces. The marker A at the sacral end is oriented towards B so that its y axis represents the direction of AB and hence the direction of the reference line. Any force which lies positioned on the y axis of the marker A beyond the location of B, or beneath A is considered to be outside the arch boundaries. This is illustrated in Figure 3-9. These forces were not included in the calculation of the compression force vector for each segment of the thrustline.
3.5.3.3.2 Forces At the Upper Arch Boundary

For the spine to be represented as an arch structure, a compressive thrust must be applied at both ends of the spine. At the upper end of the spine the centre of gravity of the head is positioned superior and anterior to the C1 vertebra. The force due to the weight of the head therefore acts outside the arch limits. However, the line of action of this force is such that the weight of the head acts downwards and provides the necessary compressive thrust for the arch.

Figure 3-9: Illustration of the limits of the spinal arch.
Equilibrium of the head about the spine requires posterior muscle and ligament forces. Those with upper attachments to the head and lower attachments to the cervical spine pull the head towards their lower attachment on the cervical spine. The forces generated at the upper attachment of the muscle therefore act above the end of the arch, but contribute to the compressive thrust on the spine. Figure 3-10 illustrates the position of the points of attachment of these forces superior to the upper arch limit.

![Diagram showing forces](image)

**Figure 3-10: Illustration of points of application of forces due to head weight and neck muscles above the upper arch limit.**

To test the effects of each posterior neck muscle group on the predicted thrustline due to body weight alone the resultant force provided by the weight of the head and the muscle force was considered to act at B providing the necessary compressive thrust (Figure 3-11) and the moments generated were included in the end support reaction calculations.
3.5.3.3.3 Forces At the Lower Arch Boundary

In this model the pelvis is considered a fixed base. Contraction of a muscle force attached between the spine and the pelvis pulls the vertebrae towards the pelvis, but does not pull the pelvis towards the spine (because it is fixed by muscles of the lower limbs). The very small amount of movement allowed at the iliosacral joint implies that the sacrum can also be considered fixed. In contrast, the head, ribcage and shoulder complex are considered moveable structures in the model, whereby when the spine moves these structures also move. Therefore only the force applied at the attachment to the vertebra is considered for the muscles which attach at the other end to the pelvis or sacrum.
The superior surface of the sacrum marks the lower boundary of the spinal arch. The reaction force generated in response to forces acting down upon it provides the necessary compressive thrust at this end. For those muscles attaching to the sacrum and pelvis, the upward forces generated at these attachments were not included in the model. (Were movement of the pelvis or sacrum included, these forces would be significant).

However, some muscle and ligament forces attaching to the L5 spinous process were also outside the arch limits when the spine was close to a fully flexed configuration. This included the upward component of the interspinous and longissimus thoracis muscle forces. Due to variation between the reference line and L5 vertebra orientation, these forces were situated outside of the reference line limits and therefore were not considered to contribute to the compressive thrust of the arch. This is illustrated in Figure 3-12. Relative to the L5/S1 intervertebral joint however, these forces would affect the joint reaction. Also, contraction of these forces could initiate movement of the vertebra and cause disturbance to the whole spine. These forces were therefore considered in joint reaction and thrustline calculations.

Figure 3-12: The L5 spinous process marker positioned below the limits of the reference line.
3.5.3.4 Definition of the Applied Spinal Forces

3.5.3.4.1 Body Weight Forces

Forces due to body weight act vertically downwards regardless of the inclination of the spinal column. Relative to the global reference frame their direction was therefore described by the Euler angle (180,0,0)\(^3\).

Data regarding the magnitude and point of application of the body weight force at each vertebral level is determined in section 3.6.2.1. The point of application of each force was expressed relative to the local co-ordinate frame of the corresponding vertebra so that the force location is moved with vertebral rotation.

3.5.3.4.2 Muscle Forces

Upon contraction of a muscle, tensile forces are generated within the fascicles. At their attachment points, for equilibrium equal and opposite forces are generated. This is illustrated in Figure 3-13. The forces which act on the vertebrae or skeletal structure at the muscle points of attachment are applied in this model. Data regarding the points of application and magnitude for the muscle forces is derived in section 3.6.3.

---

\(^3\) An Euler angle \((p,q,r)\) refers to a rotation of \(p\) degrees about the Z-axis followed by a rotation of \(q\) degrees about the new X-axis and then a rotation of \(r\) degrees about the new Z-axis. In this model the vertebral length lies along the x axis and the vertebral height along the y axis.
Within the model the points of attachment of the muscles on the vertebrae, discs and other skeletal structures are represented by markers. Single representative points on the spinous processes, transverse processes etc. are used. Several muscle fascicles may therefore attach to the same point.

The point of attachment of the muscle fascicle represents the point of application for the generated force. Each force is represented by a marker located at this point, with direction defined by the marker co-ordinate axes. A separate marker is needed to define each force due to different directional properties. This involved a large number of markers, but did not challenge the capacity of the ADAMS modelling application.

The location of the force markers were defined so that their position relative to the vertebra co-ordinate axes would be maintained when the vertebrae moved.
The direction of the muscle forces was defined according to the line of action each muscle followed. Muscles attaching close to the spine and between adjacent vertebrae are assumed to pull in an approximate straight line between their points of attachment. The y co-ordinate axis of the force marker was therefore directed along a straight line towards the position of the marker representing the point of application of the opposing force for a particular muscle fascicle.

In contrast, longer span muscles were considered to be affected by curvature (Aspden (1992)). It was assumed the muscle force would be directed towards a point representing the peak of the spinal curve over which it spanned. This is illustrated in Figure 3-14. However, if the points of attachment represented the peak point of the curve, then the muscle was assumed to be unaffected by the curve and would act in a straight line between its attachment points.
Figure 3-14: Illustration of the force directions of a muscle affected by a flexed spinal curve
3.5.4 MODEL CALCULATIONS

3.5.4.1 Calculation of the End Support Reactions for an Arch Structure Using ADAMS Command Language

In this model forces are defined as positive when acting from anterior to posterior, or upwards and moments generated by these forces are defined as positive when an anticlockwise moment is generated. Figure 3-15 shows a simple example in which the applied force (F) acts with negative perpendicular components (Fr) and positive parallel components (Fh) relative to the reference line.

Figure 3-15: Illustration of the components of the end support reaction forces for a simple parabolic arch acted upon by a single force (F).
The line of action of forces $H_1$ and $H_2$ acts directly through the ends of the arch $AB$. These therefore generate zero moment about the end supports. Determination of components $H_1$ and $H_2$ is performed merely by summation of forces satisfying laws of equilibrium:

$$H_1 + H_2 = -\sum_{i=1}^{n} F_{hi}$$

where $n$ is the number of applied forces.

In contrast, determination of components $R_1$ and $R_2$ depends upon the equilibrium of forces $R_1 + R_2 = -\sum_{i=1}^{n} F_{ri}$ and also upon equilibrium of moments generated.

### 3.5.4.1.1 Perpendicular End Support Reaction Components $R_1$ and $R_2$

Calculation of the end support reaction forces for a number of applied forces requires that these forces are systematically defined. A general definition of the signs of the components of force (positive or negative) for a given force direction is illustrated in Figure 3-16.
For a force of magnitude \( F_v \), the perpendicular and parallel components are 
\[ \text{Fr} = F_v \sin \theta, \quad \text{Fh} = F_v \cos \theta \]
where \( \theta \) is the angle between the force vector and the arch reference line. However, when accounting for any force direction the positive and negative values for the force direction rendered by the trigonometric functions are different to those required. This is shown in Figure 3-17. To ensure the values returned by the function are the same as those expected, the expression for \( \text{Fr} \) is multiplied by \((-1)\). (i.e. \( \text{Fr} = F_v \sin \theta \times (-1) \))
Figure 3-17: Comparing the required values with those returned by the trigonometric functions (Trig. Fn.).

The moment arm of each force component relative to A can be represented by a distance function in ADAMS. The function $dy(\text{Fr}, A, A)$ measures the distance of the line of action of the force component Fr to A, relative to the y co-ordinate axis of A. Similarly the function $dx(\text{Fh}, B, A)$ measures the distance of the line of action of the force component Fh to B, relative to the y co-ordinate axis of A. This is illustrated in Figure 3-18.
These functions can be used to describe the moments generated by the components of force about the end supports:

<table>
<thead>
<tr>
<th></th>
<th>About A</th>
<th>About B</th>
</tr>
</thead>
<tbody>
<tr>
<td>Moment by Fr</td>
<td>$Fr \times dy(Fr, A, A) \times (-1)$</td>
<td>$Fr \times dy(Fr, B, A) \times (-1)$</td>
</tr>
<tr>
<td>Moment by Fh</td>
<td>$Fh \times dx(Fh, A, A)$</td>
<td>$Fh \times dx(Fh, B, A)$</td>
</tr>
<tr>
<td>Moment by R1</td>
<td>zero</td>
<td>$R1 \times dy(R1, B, A) \times (-1)$</td>
</tr>
<tr>
<td>Moment by R2</td>
<td>$R2 \times dy(R2, A, A) \times (-1)$</td>
<td>zero</td>
</tr>
</tbody>
</table>

The negative sign for the Fr component function is necessary to ensure compatibility with the value returned by the trigonometric functions explained in Figure 3-17.

The actual directions of R1 and R2 are not certain and need to be determined. A negative value returned will indicate the force acting in the opposite direction to that shown.
For the simple case shown in Figure 3-15, for equilibrium the sum of moments about A must be zero:

\[(Fr \times dy(Fr, A, A) \times (-1)) + (Fh \times dx(Fh, A, A)) + (R2 \times dy(R2, A, A) \times (-1)) = 0\]

and the sum of moments about B must be zero:

\[(Fr \times dy(Fr, B, A) \times (-1)) + (Fh \times dx(Fh, B, A)) + (R1 \times dy(R1, B, A) \times (-1)) = 0\]

Therefore:

\[R2 = -\left(\frac{(Fr \times dy(Fr, A, A) \times (-1)) + (Fh \times dx(Fh, A, A))}{dy(R2, A, A) \times (-1)}\right)\]  

(1)

\[R1 = -\left(\frac{(Fr \times dy(Fr, B, A) \times (-1)) + (Fh \times dx(Fh, B, A))}{dy(R1, B, A) \times (-1)}\right)\]  

(2)

and in more general terms for an arch with n forces acting

\[R1 = -\left(\frac{\sum_{i=1}^{n} (Fr_i \times dy(Fr_i, B, A) \times (-1)) + \sum_{i=1}^{n} (Fh_i \times dx(Fh_i, B, A))}{dy(R1, B, A) \times (-1)}\right)\]

\[R2 = -\left(\frac{\sum_{i=1}^{n} (Fr_i \times dy(Fr_i, A, A) \times (-1)) + \sum_{i=1}^{n} (Fh_i \times dx(Fh_i, A, A))}{dy(R2, A, A) \times (-1)}\right)\]

These algorithms are thus used to determine the end support reactions for any number of forces acting in any direction at any position relative to the reference line of the spinal arch.

In simpler terms, the expression \(dy(R2, A, A) \times (-1)\) returns a negative value so that \(R2 = [(\text{sum of moments due to applied forces})/\text{length AB}]\) and the expression \(dy(R1, B, A) \times (-1)\) returns a positive value so that \(R1 = -[\text{(sum of moments due to applied forces})/\text{length AB}]\).
3.5.4.2 Calculation of the Thrustline Co-ordinates for an Arch Structure

The predicted thrustline for the set of applied forces provides an immediate indication of the stability of the spine. Calculation of the thrustline co-ordinates in this model is based on the graphical method described by Heyman (1982) using funicular polygons. Figure 3-19 shows forces $F_1$ and $F_2$ acting on a symmetric vertical arch and the resulting end support reactions $R_1$, $R_2$, $H_1$ and $H_2$.

Figure 3-19: Illustration of external forces acting on a symmetric vertical arch.

Figure 3-20 shows the resultant compression force vectors ($C_1$, $C_2$, $C_3$) acting in each section of the arch. The angles made by each vector relative to the direction of $H_1$ (i.e. relative to the reference line) are defined as $\theta_1$, $\theta_2$ and $\theta_3$. 
Figure 3-20: A force polygon diagram for the arch structure.

The thrustline can then be constructed (Figure 3-21). The y co-ordinates of each thrustline portion correspond to the y co-ordinates of the points of force application.

Figure 3-21: The inclination of the resulting portions of the thrustline.
The angles between each portion of the thrustline and the reference line can be used to determine the x co-ordinates of the thrustline points. For the case shown in Figure 3-21, \( \tan \theta_i = \frac{y_i}{d_{yi}} \) where \( d_{y1} \) is the distance between A and \( y_1 \) and \( d_{yi} \) is the distance between \( y_{i-1} \) and \( y_i \) relative to the reference line.

Also, \( \tan \theta_1 = \frac{R_1}{H_1} \)  \( \tan \theta_i = \frac{R_1 + \sum_{k=1}^{i-1} F_{rk}}{H_1 + \sum_{k=1}^{i-1} F_{hk}} \) for \( i = 2 \) to \( n+1 \) where \( n \) is the number of applied forces.

For each section in the arch, the x co-ordinates of the thrustline can thus be calculated using the equations:

\[
x_i = \frac{R_1}{H_1} \times d_{y1}, \quad x_i = x_{i-1} + \left[ \frac{R_1 + \sum_{k=1}^{i-1} F_{rk}}{H_1 + \sum_{k=1}^{i-1} F_{hk}} \times d_{yi} \right] \quad \text{for} \ i = 2 \quad \text{to} \ n+1 \quad \text{where} \ n \quad \text{is the number of applied forces.}
\]

However, in some cases the x values calculated do not return the thrustline to the original reference line (Figure 3-22). In these cases, a shear transformation must be used to return the line AB' and the consequent points of the thrustline to the line AB. The shear transformation is \( x' = x + a \times y_i \) where the shear coefficient \( a = -\frac{x_n}{y_n} \).
The $x$ and $y$ co-ordinates of the predicted thrustline for an arch under the application of external forces can thus be determined using the algorithms derived.

3.5.4.2.1 Sorting Procedure

The start of each section of the thrustline follows on from the end of the previous portion and is therefore dependent upon the $x$ co-ordinates of the previous sections. The nature of the calculations therefore depends upon forces acting sequentially along the length of the arch. This requires points of application of the forces relative to the reference line to be sorted by increasing distance from $A$. 
The labelling system used in the ADAMS macros to associate variables with the correct vertebral level is also used to ensure the sequential order of the applied forces. The control variable ‘numberofforces’ is assigned a value according to the number of applied forces to be tested. For bodyweight alone, the variable value is 24, so that the force markers are labelled from 1 to 24. These are initially written to correspond to the levels of the spine and do therefore not require sorting. However, the subsequent addition of a muscle group results in forces labelled from 25 onwards with possible locations between the bodyweight forces. For each force marker, there are also three variables: \( f_v \) representing the magnitude of the force, \( f_r \) representing the component of force perpendicular to the reference line and \( f_h \) representing the component of force parallel to the reference line. Each of these shares the same suffix as their corresponding force marker.

A sort procedure must therefore compare each pair of force markers with consecutive labels for their y co-ordinate relative to AB. An example is shown in Figure 3-23 for six force markers labelled according to their sequential position. If the lower force marker has a higher suffix value, this value and the suffix values for the associated force magnitude and component variables are exchanged.
3.5.4.2.2 Equal Locations

The algorithm also requires that no two force markers share the same y location. If the two markers were in the same location, the value $dy_i$ in

$$x_i = x_{i-1} + \left[ \frac{R1 + \sum_{i=1}^{k} Fr_k}{H1 + \sum_{i=1}^{k} Fh_k} \right] \times dy_i$$

between them would be zero and would result in the same increment of thrustline co-ordinate as the previous, regardless of the applied force value. The forces acting at the same point were therefore resolved into a single force, ensuring equilibrium of forces and moments is maintained.
3.5.4.3 Graphical Display of the Predicted Thrustline

The algorithms derived determine the x and y co-ordinates of each section of the thrustline. Representation of the thrustline graphically in ADAMS can be achieved by constructing straight lines between each thrustline co-ordinate set. Using an 'outline' geometry requires these lines be constructed between markers located at these co-ordinates. The initial and final markers for the sheared thrustline coincide with the locations of A and B.

3.5.4.4 Calculation of the Joint Reactions at each Intervertebral Level Using ADAMS Command Language

The compressive thrust at each level of the spine represents the resultant force at that level. Expressing this force in terms of shear (perpendicular) and compression (parallel) components relative to the intervertebral joint axis thus represents the total load at each intervertebral level. (It is assumed the joint axis follows the same direction as the axis of the vertebra above). For equilibrium of a single section of an arch, the intervertebral joint provides a reaction force which is equal and opposite to this resultant force.

Calculation of the load at each intervertebral joint was performed by summing all the forces acting above the joint level in terms of the perpendicular (totalFr) and parallel (totalFh) components relative to the reference line. The total magnitude of the resultant force (TotalF) was then determined based on Pythagoras' theorem as the square root of the sum of squares of its components.
The direction of this force was defined by the function: \[180 \times (\text{totalFh} < 0) - \tan(\text{totalFr} / \text{totalFh})\]. The boolean expression \((\text{fh}<0)\) returns a value of 1 if true in ADAMS. For a force acting between 90 and 180 degrees shown in Figure 3-24 the expression \(\tan(\text{totalFr}/\text{totalFh})\) returns a positive value and \(\text{Fh}<0\) returns a value of 1. The angle relative to AB is therefore \((180-\theta)\).

![Figure 3-24: Illustration of calculation of joint reactions.](image)

The magnitude of the total force (\(\text{totalF}\)) relative to the joint axis (\(\text{totalR}\)) was then determined using the angle between the reference line and the joint y coordinate axis. The components of force relative to the joint axis in terms of shear (x axis) and compression (y axis) were then determined using the relations:

\[
\text{Compression} = \text{totalR} \cos \phi, \quad \text{Shear} = \text{totalR} \sin \phi \quad \text{where} \quad \phi \quad \text{is the angle between the resultant and the joint axis.}
\]

The joint reaction force components for equilibrium are of equal magnitude and opposite direction to these force components.
These calculations can be summarised in five steps:

1) Summing of all the forces acting above the joint level in terms of the perpendicular (totalFr) and parallel (totalFh) components relative to the reference line.
2) Determination of resultant force (totalF) magnitude and direction $\theta$ relative to reference line.
3) Determination of resultant force magnitude (totalR) and direction $\phi$ relative to joint axes.
4) Determination of magnitude of shear and compression components.
5) Determination of joint reaction force components of equal magnitude and opposite direction to these force components.

3.5.4.5 Output of Joint Reactions, Thrustline Co-ordinates and Muscle Forces

The stability of the spine is indicated by the position of the predicted thrustline relative to the path of the vertebral bodies. Output of the thrustline co-ordinates and the vertebral body centre, anterior and posterior length co-ordinates thus provides quantitative information regarding the stability of the spine. Also, the calculated joint reactions provides information regarding the loading due to body weight, external loading, muscle and ligament forces at each level of the spine.

Output of these variables requires the ADAMS command 'list_info'. The use of wild cards in ADAMS command language such as "fm*" allows data for complete set of force markers with prefix 'fm' to be listed into one file.

The output is presented in a combined form of alphanumeric digits which cannot be automatically processed in spreadsheets or other files.
To be able to read this automatically into spreadsheets or other files required the numeric form of this data be processed into manageable form. A Perl program was therefore written for this process. The output of the program was stored in a formatted file as tabulated data, with each row containing the numbers of the variables and the corresponding variable values. Each value belonged to a column, with the variable names the titles of each column.

As an example the output for variable fr3:

Object Name: spine.fr3
Object Type: Variable
Parent Type: Model
Real Value(s): -0.3896454715 NO UNITS
Units: no_units
Comment String: None

was processed into tabular format:

<table>
<thead>
<tr>
<th>Variable Number</th>
<th>Variable Fr</th>
<th>Variable Fh</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 value</td>
<td>value</td>
<td>value</td>
</tr>
<tr>
<td>2 value</td>
<td>value</td>
<td>value</td>
</tr>
<tr>
<td>3 -0.3896454715</td>
<td>value</td>
<td>value</td>
</tr>
</tbody>
</table>

3.5.5 SUMMARY

Based upon the input data for the vertebral configuration and applied forces, the model predicts the resulting thrustlines and joint reactions. The user may specify the vertebral dimensions or angles, the body weight data, and the muscle groups to be investigated. Algorithms within the model define the relation between vertebral configuration and the applied forces. The path of the resultant force at each level of the spine is then determined. Collectively each force direction is represented by the overall path of the thrustline which can be displayed graphically when the required co-ordinates are calculated. The consequent joint reaction forces at each vertebral level are then determined. Values for the input data regarding vertebral configuration, body weight and muscle forces are discussed in the following sections.
3.6 DERIVATION OF INPUT DATA FOR THE MODEL

3.6.1 SIMULATION OF VERTEBRAL CONFIGURATION FOR VARIOUS POSTURES

Input data for the model required values for the vertebral and disc length and height and the relative angles of orientation between each vertebra. However, existing data regarding the configuration of the vertebrae in various postures is very limited, due to the large variation in dimensions among specimens and because the complex movement of the spine allows a large range of postures.

3.6.1.1 Determination of Vertebral and Disc Dimensions and Orientations.

Several geometric studies regarding measurements of vertebra and disc dimensions are available in the literature (Lanier, 1939, Gilad and Nissan, 1984, Panjabi et al., 1991a and b and 1992 and Stepney et al, 1996). The in-house measurements of Stepney et al. (1996) were considered appropriate, so that compatibility between the mathematical model investigations and the geometric model to be overlaid could be maintained (see page 146, section 3.3.6).
Also, the study of Stepney et al. (1996) provided data for parameters of the vertebral body, spinous and transverse processes, superior and inferior facets, spinal canal and pedicles for muscle attachment. These were considered simpler than the complex description of surfaces for the facets reported by Panjabi et al. (1991a and b). This data was therefore considered more appropriate for the spinal model in which muscle attachments are to be described as simple points. Geometric properties of the vertebrae and discs included in the model were therefore included based on the work of Stepney et al. (1996).

In the study of Stepney et al. (1996) measurements for the vertebral body length were taken for the superior, middle and inferior dimensions. Due to the circumferential concavity, the middle values were less than those reported for the superior and inferior dimensions. Consequently, to obtain a length value which would represent the approximate dimensions of the vertebral body, represented as a regular box in ADAMS, the average of the superior and inferior length values was used. For the vertebral height, only one representative value was reported. The difference in anterior and posterior height for the vertebra were included in measurements made for the inter vertebral disc above.

The data reported by Stepney et al. (1996) did not however include dimensions for the axis or atlas vertebra. Scaled cross sectional anatomy drawings (Eccleshymeyer and Shoemaker, 1911) were used to obtain scaled measurements which corresponded to 40mm and 44mm for the actual lengths of the atlas and axis vertebrae respectively. Although the atlas does not have a distinct vertebral body, it was assumed the origin of the atlas would coincide with the centre of the dens protruding through it. This is illustrated in Figure 3-25.
The height of the axis vertebral body and dens was estimated as 36mm from the scaled cross section anatomy drawings. This was in close agreement with the measured value for the axis by Lanier (1939) of 37.185mm. It was assumed that the dens projected through to the top of the atlas, so that this height represented the height of both the axis and atlas vertebrae. Two boxes representing the vertebral body of the axis and the equivalent cavity occupied by the dens in the atlas were therefore produced with equal height and length dimensions. Using the scaled cross section anatomy drawings, the vertebral body of the axis was approximately three tenths of the total vertebra length. The height and length dimensions of each of the representative boxes were therefore 18mm and 12mm respectively.
3.6.1.2 Configuration in the Erect Posture

The variation in disc and vertebra dimensions for each individual at every level of the spine results in numerous possible configurations. An 'ideal' erect posture is described by Yoganandan et al. (1987). They report "the centre line of gravity of the spinal column passes through the dens of the axis, T2 and T12 vertebral bodies and the promontory of the sacrum". Thus, a line drawn from the dens of the axis to the promontory of the sacrum for the spine would be expected to be vertical (Figure 3-26). An erect configuration of the spine which matches these criteria was therefore sought.
Figure 3-26: Showing the line of gravity for the spinal column.  
(Adapted from Yoganandan et al., 1987)
In the study by Stepney et al. (1996) the orientation of the vertebra above relative to the vertebra below was determined by the overall difference in height between the anterior and posterior height for vertebra and disc combined. However, data for the cervical region did not match the required configuration, possibly due to the small dimensions of the discs and vertebrae. Measurements are difficult in this region with a small variation having a greater cumulative effect. The angles of orientation used were therefore modified slightly until the resulting configuration showed a fair agreement with the configuration shown by Yoganandan et al. (1987). The values are shown in the appendix (section 8.1) and the resulting configuration shown in Figure 3-27.

Figure 3-27: The configuration of the spine in an approximate erect posture.
3.6.1.3 Configuration in the Flexed Postures

Due to lack of available data describing the configuration of the spine in flexion, postures were simulated based upon ergonomic textbook pictures (Helander, 1995). Three positions representing various degrees of flexion during lifting were investigated. The posture shown in Figure 3-28 was classified as a slightly flexed (flex1) posture, in Figure 3-29 as a more flexed posture (flex2) and in Figure 3-30 as a near fully flexed (flex3) posture.

![Figure 3-28: The spine in a slightly flexed posture (flex1).
(Taken from Helander, 1995.)](image)
Figure 3-29: The spine in a more flexed posture (flex2).
(Taken from Helander, 1995.)

Figure 3-30: The spine in a near fully flexed posture (flex3).
(Taken from Helander, 1995.)
Again, as for the erect posture, angles of orientation were approximated until a close agreement between the configurations had been obtained. The angles of orientation for each of the three postures are shown in the appendix (section 8.1) and the resulting configurations of the spine are shown below in Figure 3-31 for the flex1 posture, Figure 3-32 for the flex2 posture and Figure 3-33 for the flex3 posture.

Figure 3-31: Illustration of the configuration of the spine for the flex1 posture.
To ensure the angles of orientation lie within physiological ranges, the changes in angle relative to the erect posture were compared against reported ranges of flexion-extension compiled from the literature (White and Panjabi, 1990). The values are reported in the appendix (section 8.1) and were within the required range.
3.6.2 A DISTRIBUTED LOADING PATTERN FOR BODY WEIGHT ALONG THE SPINE AND THE ADDITION OF EXTERNAL LOADS

3.6.2.1 Derivation of a Force Distribution Pattern for Body Weight

Data describing the individual body weight forces at each level of the spine was obtained from studies of the thoraco-lumbar spine (Takashima et al., 1979) and of the cervical spine (Merril et al., 1984) available in the literature. This data is used to derive a loading pattern for the upper body weight to be applied in the whole spine model.

The force distribution pattern reported by Takashima et al. (1979) considered the weight of an imaginary trunk slice at each vertebral level of the thoraco-lumbar spine. The weight of each trunk slice was assumed to act through the centroid of the slice, at a certain distance from the centre of the vertebral body (eccentricity). Each imaginary slice was assumed to be taken for the spine in the upright posture parallel to the horizontal plane at the base of each vertebra (Figure 3-34). The volume and centroid of each slice were estimated using cross sectional anatomy drawings (Eccleshymer and Shoemaker, 1911), scaled to agree with the gross trunk dimensions reported by Clauser et al. (1969) whose data was based upon a larger population sample. The mass and then the weight of each trunk slice was then determined assuming homogeneity and a density value of 1.019 g/cm³.
Takashima et al. (1979) have also reported the weight of the arms to be 18 N acting on each of T2, T3 and T4 vertebra, acting 3 cm posterior to the vertebral centres. The force due to the mass of the arms at the T2 level and the mass of the trunk at this level can be represented by a single resultant force which is determined by equilibrium of forces and moments. The resultant forces were also calculated for T3 and T4 levels using this method. One force representing the combined upper body weight and arm weight can therefore be applied at each of these levels. The values for the weight and eccentricity at each level are shown in Table 3-1.
Table 3-1: Showing the weight of the upper body acting and its eccentricity relative to each vertebral level of the thoraco-lumbar spine derived from data of Takashima et al. (1979) for the erect posture.

<table>
<thead>
<tr>
<th>Vertebra Level</th>
<th>Weight (N)</th>
<th>Eccentricity (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1</td>
<td>27</td>
<td>-2.6</td>
</tr>
<tr>
<td>T2</td>
<td>31</td>
<td>-1.2378</td>
</tr>
<tr>
<td>T3</td>
<td>27</td>
<td>-1.8</td>
</tr>
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<td>T4</td>
<td>31</td>
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</tr>
<tr>
<td>T5</td>
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<td>3.7</td>
</tr>
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<td>T6</td>
<td>11</td>
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</tr>
<tr>
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<td>T8</td>
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<td>T9</td>
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</tr>
<tr>
<td>T10</td>
<td>18</td>
<td>6.2</td>
</tr>
<tr>
<td>T11</td>
<td>18</td>
<td>6.3</td>
</tr>
<tr>
<td>T12</td>
<td>19</td>
<td>6.0</td>
</tr>
<tr>
<td>L1</td>
<td>21</td>
<td>5.3</td>
</tr>
<tr>
<td>L2</td>
<td>19</td>
<td>4.4</td>
</tr>
<tr>
<td>L3</td>
<td>19</td>
<td>2.8</td>
</tr>
<tr>
<td>L4</td>
<td>25</td>
<td>3.4</td>
</tr>
<tr>
<td>L5</td>
<td>18</td>
<td>4.3</td>
</tr>
<tr>
<td>SACRUM</td>
<td>7</td>
<td>6.7</td>
</tr>
</tbody>
</table>

In the study by Merrill et al. (1984), the mass values at each cervical vertebra level were determined from slices of cadaver specimens. The centre of mass of the neck at each vertebral level was assumed to correspond to the location of the spinal cord (assumed here to be the centre of the spinal canal).

The distance between the centre of the spinal canal and the vertebral body centre was calculated using geometric data for the vertebrae C3-C7 reported by Stepney et al. (1996) and scaled cross section drawings (Ecdeshymer and Shoemaker, 1911) for C1 and C2. These distances and the weight values are shown in Table 3-2. The difference between these distances and the eccentricity values determined relative to a horizontal slice in the upright posture by Takashima et al. (1979) for the thoraco-lumbar regions is illustrated in Figure 3-35. The equivalent distances relative to an imaginary horizontal slice (eccentricity values) for the geometry used are also shown in Table 3-2.
In the model by Merril et al. (1984) the skull mass is 3.53 Kg applied 1.94 cm anterior and 4.73 cm superior to the centre of the C1 vertebra, expressed in global co-ordinates. For the erect posture used in our study this is equivalent to 0.45 cm anterior and 4.55 cm superior to the vertebral body centre.

Table 3-2: Showing the weight of the upper body acting relative to each cervical vertebral level derived from data of Merrill et al. (1984).

<table>
<thead>
<tr>
<th>Vertebra Level</th>
<th>Weight (N)</th>
<th>Distances from Vertebral Centre to Vertebral body Centre (cm)</th>
<th>Eccentricity (horizontal distance) (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C1</td>
<td>1.5288</td>
<td>-1.5</td>
<td>-1.49</td>
</tr>
<tr>
<td>C2</td>
<td>1.5288</td>
<td>-1.5</td>
<td>-1.49</td>
</tr>
<tr>
<td>C3</td>
<td>1.5288</td>
<td>-0.927</td>
<td>-0.91</td>
</tr>
<tr>
<td>C4</td>
<td>2.009</td>
<td>-0.983</td>
<td>-0.98</td>
</tr>
<tr>
<td>C5</td>
<td>2.6362</td>
<td>-1.052</td>
<td>-1.05</td>
</tr>
<tr>
<td>C6</td>
<td>2.2148</td>
<td>-1.1075</td>
<td>-1.10</td>
</tr>
<tr>
<td>C7</td>
<td>3.920</td>
<td>-1.187</td>
<td>-1.16</td>
</tr>
</tbody>
</table>

Figure 3-35: Illustration of the distances measured for the forces due to body weight acting at the cervical vertebral levels.
When testing for each configuration of the spine, the points of application of these forces due to body weight require positioning relative to the co-ordinate system of each vertebra at the corresponding level. This ensures that as the spinal configuration is altered, the weight positions relative to the vertebral body centres are maintained. This was considered when describing these forces in the computer based modelling application.

The derived distributed force patterns for body weight for each posture are illustrated in chapter 4 (pages 247-249) in which the model predictions using these patterns are reported.
3.6.2.2 Additional External Loads Carried in the Arms

When lifting an external load, the load held in the hands can be represented simply as an extension to the arm segments. The weight of the arms is therefore increased. In Aspden's arch model (e.g. Aspden, 1989), the force due to the external load lifted was assumed to act at the T6 vertebra. However, using the data of Takashima et al. (1979), the weight of the arms acting on the spine is distributed between the T2, T3 and T4 vertebra levels. The effects of an external load to the spine can therefore be considered through additional force at these three vertebral levels (Figure 3-36).

Figure 3-36: Illustration of the weights of the arms and an external load represented collectively, acting at T2, T3 and T4 levels.
In the study by Takashima et al. (1979), the arms are assumed to be hanging freely vertically downwards by the sides of the body. The weight of the arms is reported to act posteriorly to the T2, T3 and T4 vertebral body centres. This is due to the positioning of the shoulder complex posterior to the spine. However, movement of the arms forwards to lift a load moves the position of the centre of mass of the arms and shoulders anteriorly. However, details of the precise measurement of this movement is not available to calculate new eccentricity positions. An approximation of these positions was therefore to assume the weight of the arms and external load acted at the vertebral centres. The additional load of 900N lifted in the hands which was simulated in Aspden's previous studies can be applied to the new spinal model as forces of 300N each acting at T2, T3 and T4 levels.
3.6.3 DERIVATION OF A DISTRIBUTED PATTERN OF MUSCLE FORCES ACTING ALONG THE SPINE

3.6.3.1 Direct and Indirect Muscle Forces Acting on the Spine

Muscle groups which regulate the behaviour of the spine can be categorised into two distinct groups:

1) Direct: Muscle groups which attach directly to the spine at one or both ends

2) Indirect Forces: Muscle groups which attach to the head, ribcage or pelvis and do not attach to the spine but cause the spine to bend, thereby generating moments about the spine.

For each of these muscle groups, a complete description requires definition of the points of application, line of action and magnitude of each force.
3.6.3.1.1 Direct Forces

Anatomical textbooks and various studies from the literature reveal the muscle groups with distinct attachments to the spine. These muscles can be further classified into muscle groups which attach at both ends to the spine (internal) and muscle groups which attach at one end to the spine and at the other end to structures such as the head, ribcage, shoulder complex or pelvis (internal-external). The muscles for each category and the levels over which they act are shown in Figure 3-37 and Figure 3-38. Although quadratus lumborum has one fascicle which attaches to the ribcage and pelvis, the remaining fascicles attach to the vertebrae. The muscle was therefore classified as direct.
Figure 3-37: Points of attachment for internal direct muscle groups of the spine. (Sources of reference are listed on page 208.)
Figure 3-38: Points of attachment for internal-external direct muscle groups of the spine. (Sources of reference are listed on page 208.)
3.6.3.1.2 Indirect Forces

Muscle groups which were identified to generate significant bending moments on the spine in the sagittal plane are listed in Table 3-3.

**Table 3-3: Listing the muscle groups with indirect action on the spine included in investigations.**

<table>
<thead>
<tr>
<th>Muscle Group</th>
</tr>
</thead>
<tbody>
<tr>
<td>rectus abdominis</td>
</tr>
<tr>
<td>internal oblique</td>
</tr>
<tr>
<td>external oblique</td>
</tr>
<tr>
<td>iliocostalis lumborum pars thoracis</td>
</tr>
<tr>
<td>sternocleidomastoid</td>
</tr>
</tbody>
</table>
3.6.3.2 Selection of Muscle Groups for Investigating Spinal Stability

Due to the immense number of muscle groups acting on the spine, it was necessary to consider only those which were likely to contribute most to the stability of the spine in the sagittal plane. Based upon the muscle maps (Figure 3-37 and Figure 3-38) and guided by descriptions of the role of these muscles in the literature and the anatomic textbooks, these muscle groups were identified.

Muscles excluded from consideration were:

i) for elevating the ribs: levatores costarum
serratus posterior superior
serratus posterior inferior

ii) for control of the shoulder complex: rhomboideus major and minor
trapezius
latissimus dorsi (upper limb)
levator scapulae
In contrast, muscles which were considered important for stability in the sagittal plane were:

i) for extending the spine in the sagittal plane:
   longissimus thoracis pars thoracis
   and lumborum
   iliocostalis lumborum pars
   lumborum  (lumbar iliocostalis)
   spinalis and semispinalis thoracis
   longissimus cervicis
   quadratus lumborum (pelvis fixed)
   lumbar multifidus
   splenius cervicis
   spinalis cervicis,
   iliocostalis cervicis

ii) for extending the head in the sagittal plane:
   longissimus capitis
   splenius capitis
   spinalis capitis
   obliquus capitis superior
   rectus capitis posterior major
   rectus capitis posterior minor

iii) for flexion of the cervical spine in the sagittal plane:
   longus coli
   scalenus (ant. med. and post.)

Due to the similar attachments of the scalene muscles in the sagittal plane, the posterior muscle was selected as a representative group to be studied.
vi) "postural muscles, necessary for keeping the spine together":
   rotatores
   interspinalis
   intertransverse
   multifidus

iii) for equilibrium of the head in the frontal plane:
   obliquus capitis inferior
   rectus capitis lateralis

iv) for flexion of the head in the sagittal plane:
   longus capitis
   rectus capitis anterior

Finally, some indirect muscles which generated bending moments about the spine but did not attach directly to it were considered.

v) for extension of the spine
   iliocostalis lumborum pars thoracis
   (thoracic iliocostalis)

vi) due to the report of coactivation of the abdominal muscles during lifting activities, the roles of rectus abdominis, internal and external oblique and transverse abdominis were considered.

*Gray's Anatomy, 1967*
3.6.3.3 Points of Application of Direct Muscle Forces

The points of attachment of the muscle fascicles to the spine represent the points of application of the forces which they generate. Descriptions from anatomical textbooks are relatively simple, defining the major sites of attachment for each muscle group. In contrast, various studies performed for the lumbar and cervical muscle groups report the attachments of each individual muscle fascicle. However, such studies have not been performed for all spinal muscle groups. Therefore where possible a detailed description of the points of attachment of the individual fascicles using the literature was included and for those muscles not studied in detail, a simpler description from anatomic textbooks for those muscles was used.

The detailed studies investigating the anatomy of individual muscle fascicles are based upon cadaver studies. They include:

- psoas major (Bodguk et al., 1992)
- lumbar multifidus (Macintosh and Bogduk, 1986)
- erector spinae (longissimus thoracis and iliocostalis lumborum) (Macintosh and Bogduk, 1987, 1991)
- rhomboideus (Johnson et al., 1996)
- levator scapula (Johnson et al., 1994)
- trapezius (Johnson et al., 1994)
- spinalis capitis (Martin, 1994)
For other muscle groups three sources of textbooks were used:

2) The Head, Neck and Trunk (Warfel, 1985)
3) Gray's Anatomy (1967)

Figure 3-39 summarises the points of attachment for the muscle groups considered.

![Diagram of muscle groups and their attachments](image-url)

Figure 3-39: Showing the points of attachment of the spinal muscle groups considered. (Sources of reference are listed on page 208.)
3.6.3.4 Points of application of indirect forces

The detailed study performed by Macintosh and Bogduk (1987) describes the points of attachment of the individual fascicles of thoracic iliocostalis on the ribcage and pelvis.

In contrast, anatomic textbook descriptions report the abdominal muscles to be broad sheets of muscle, from which the precise points of application of the generated forces is difficult to determine.

The anatomic study by Dumas et al. (1991) describes points of attachment and lines of action for the forces for the abdominal muscles rectus abdominis, transverse abdominis and the internal and external obliques. The action of these muscles are represented at most by four forces whose points of application cannot possibly represent the entire area over which the muscle acts. However, while these may not match the detail of the study performed by Macintosh and Bogduk (1987) they can be used to investigate the representative action of these muscle forces.

Figure 3-40 illustrates the lines of action and points of application for these forces as reported by Dumas et al. (1991). In contrast, illustrations in Gray’s Anatomy show the orientation of the rectus abdominis muscle to be approximately parallel to the longitudinal axis of the spine in the upright posture.
3.6.3.5 Representation of the Points of Muscle Attachment

3.6.3.5.1 Muscle Attachment to the Vertebrae

Although the points of attachment of each muscle group varies on the surfaces of the vertebral processes, a single representative point was used for each. These are illustrated in Figure 3-41 and were:

- spinous process (e.g. for interspinalis, spinalis thoracis)
- transverse process (e.g. for longissimus thoracis, intertransverse)
- articular processes (e.g. for semispinalis capitis in cervical region)
- vertebral lamina (for rotatores in the thoracic region)
- mamillary processes (for multifidus in the lumbar region)
The x and y co-ordinates of these representative points were determined using the parameters of Stepney et al. (1996), described in the appendix (section 8.2).

![Diagram of vertebra with labeled points](image)

**Figure 3-41: Representative points for muscle attachment on the vertebra.**

### 3.6.3.5.2 Muscle Attachments to the Pelvis and Sacrum

The detailed studies of the lumbar muscle groups (page 208) provide illustrations of their points of attachment on the sacrum and pelvis. A mapping from the illustrations in the studies to the model geometry was made so that the approximate location of the muscle attachments on the sacrum and pelvis could be represented. This involved determining the distance of these points relative to the superior surface of the sacrum and scaling these values so that their positions relative to the model origin could be obtained.
Similarly Bogduk et al. (1992) report the fascicles of psoas merge into a common tendon which attaches to the trochanter (Figure 3-42). The approximate position at which these fascicles merged was identified and again expressed relative to the superior surface of the sacrum on the scale of the model.

Figure 3-42: Illustration of the points of attachment of psoas muscle fascicles to a common tendon. (Adapted from Bogduk et al, 1992.)

3.6.3.5.3 Muscle Attachments to the Head

The points of attachment of the neck muscles required definition of certain points on the skull and on the atlas and axis vertebrae.
3.6.3.5.3.1 Atlas and Axis Vertebra

For certain muscle groups it was necessary to determine representative points of attachment on the atlas and axis. These were:

i) axis spinous process: e.g. for attachment of rectus capitis posterior major.

ii) atlas transverse processes: e.g. for attachment of rectus capitis lateralis

iii) atlas lateral mass: e.g. for attachment of rectus capitis anterior

iv) atlas posterior tubercles: e.g. for attachment of rectus capitis posterior minor.

These were estimated using the scaled cross sectional anatomy drawings of Eccleshymer and Shoemaker (1911) and are summarised in Table 3-4.

Table 3-4: Approximate positions of muscle points of attachment on the axis and atlas.

<table>
<thead>
<tr>
<th>Vertebral Feature</th>
<th>Distance from vertebral origin</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X-axis</td>
</tr>
<tr>
<td>Atlas posterior tubercle</td>
<td>$\frac{5}{6}$ * representative vertebral body length</td>
</tr>
<tr>
<td>Axis spinous process</td>
<td>$\frac{2}{3}$ * vertebral body length</td>
</tr>
<tr>
<td>Atlas lateral mass</td>
<td>Level with origin</td>
</tr>
<tr>
<td>Atlas transverse process</td>
<td>Level with origin</td>
</tr>
</tbody>
</table>
For the attachment of rectus capitis anterior, it was assumed the lateral mass of the atlas was approximately level with the origin of the atlas in the sagittal plane and at approximately mid height. This was therefore positioned at (0, (vbheight/2),0) relative to the origin. For attachment of rectus capitis lateralis the transverse process of the atlas was positioned at the centre of the superior surface relative to the atlas origin.

3.6.3.5.3.2 Attachments to the Skull

3.6.3.5.3.2.1 X Co-ordinates in the Sagittal Plane

A cross sectional illustration of the head at the level of the occipital condyles from Gray's Anatomy (1967) provided an indication of the relative positions of these muscle groups in the sagittal plane at this level (Figure 3-43). The scaled cross sectional anatomy drawings of Eccleshymer and Shoemaker (1911) were then used to obtain actual size measurements relative to the C1 vertebra. Calculations are provided in the appendix (section 8.3).
Figure 3-43: A cross sectional illustration of the head at the level of the occipital condyles. (Adapted from Gray's Anatomy, 1967.)

3.6.3.5.3.2.2 Y Co-ordinates in the Sagittal Plane

Illustrations in Gray's anatomy (1967) were also used to identify the relative positions of the muscle points of attachment in the vertical plane. For simplification the approximate locations of the jugular process, mastoid process and basilar part of the occipital bone were assumed to be approximately at the same level (illustrated in 3-44). The points of attachment for rectus capitis lateralis, rectus capitis anterior, splenius capitis, longissimus capitis and longus capitis were thus assumed to have the same y co-ordinate.
The rectus capitis posterior major and minor and obliquisus capitis superior were assumed to attach at the approximate level of the inferior nuchal line. The y co-ordinate midway between the levels of the superior and inferior nuchal lines was used to represent the semispinalis capitis attachment. The spinalis capitis muscle was positioned at the y co-ordinate midway between semispinalis capitis and rectus capitis posterior minor attachment points (Martin, 1994).

Again scaled cross sectional anatomy drawings (Eccleshymer and Shoemaker, 1911) were then used to obtain actual size measurements relative to the C1 vertebra and values are reported in the appendix (section 8.3).

3-44: A lateral view illustration of the head.
(Adapted from Gray's Anatomy, 1967.)
3.6.3.5.4 Muscle Attachments to the Rib Cage

The attachments of iliocostalis cervicis at the rib angles and of the scalene muscles to the scalene tubercles were considered to be located approximately in the same position as the transverse processes of the vertebrae at that level in the sagittal plane.

Similarly the attachments of the thoracic iliocostalis to the ribs was assumed to be approximated by the position of the transverse processes of the vertebrae at the equivalent level.

In contrast, the large abdominal muscles which enclosed the abdominal cavity attached to the ribs at a considerable distance anterior to the vertebrae. Cross sectional anatomy drawings were used to approximate this distance.

The attachment of rectus abdominis to the ribs is approximately level with the T11, T10, T9 and T8 vertebrae, at a distance of approximately 145mm anterior. The attachments of the internal and external obliques are approximately level with T12, T11 and T10 (internal oblique) and also at L1 (external oblique). Using cross sectional anatomy drawings those at T11 and T12 were approximately 87mm and 80mm anterior to the corresponding vertebrae. Based on the illustrations of Dumas et al. (1991) (page 231) muscle attachments move closer to the vertebrae at lower spinal levels. Therefore at T12 and L1, distances were approximated as 72.5mm and 65mm.
For the transverse abdominis, the attachment to the linea alba was considered. Using cross sectional anatomy drawings, the attachment of this muscle was approximately 150mm anterior to the T12 vertebra. From the attachments at the L1, L2 and L3 spinous processes, imaginary lines were extended according to the angle of orientation (reported by Dumas et al., 1991) until the distance anterior to the T12 vertebra had been reached. The equivalent levels of vertebral attachment were then determined as L3, L4 and L5.

3.6.3.6 Lines of Action of the Muscle Forces

The direction of the applied forces depends upon the line of action of the muscle force. Muscles attaching close to the spine and between adjacent vertebrae are assumed to pull in an approximate straight line between their points of attachment.

In contrast, more superficial muscles which are also of greater length are affected by the curvature of the spine. For example the thoracic fibres of longissimus thoracis which span twelve vertebral levels follow a curved path corresponding to the curvature of the thoracic spine. The line of action of these muscles is therefore directly affected by spinal curvature.
Muscles which spanned more than two segments of the thoracic region were thought to be affected by the curvature:

- longissimus thoracis
- thoracic iliocostalis
- longissimus cervicis
- semispinalis cervicis
- splenius cervicis
- spinalis cervicis
- spinalis thoracis
- semispinalis thoracis

Also, the vertical portion of longus coli which spans the cervical region was thought to be affected by cervical lordosis.
3.6.3.7 Determination of the Magnitude of the Muscle Forces Acting on the Spine

For each muscle group, an indication of the maximum force which can be generated was necessary so that the full load bearing capacity of the spine could be investigated. The relation between the maximum force (N) generated by a muscle and its physiologic cross sectional area (cm²) was used:

\[ \text{force}_{\text{max}} = \text{PCSA}_{\text{max}} \times k \]

where \( k \) is a force coefficient indicating the maximum muscle force generated per unit of cross sectional area. The value for \( k \) in the literature has varied from 30-100 N/cm² (McGill and Norman, 1987). Using this force prediction method implies the force for every muscle whose cross sectional area is known can be estimated. However, the amount of data available regarding the cross sectional areas of the spinal muscles is limiting.

3.6.3.7.1 Data Available in the Literature for the Cross Sectional Areas of the Spinal Muscles

Studies found in the literature do not include every spinal muscle group. Also, the level of detail to which each muscle group has been studied and the techniques used have varied. Some studies have been based upon cadaver material and others upon living tissue.

Studies reporting the greatest detail regarding the force generating capacity are for the lumbar region muscles. These have been instigated by the requirement for accurate input data into the developing low back models.
3.6.3.7.1.1 Lumbar Muscles

Study of the trunk muscles at the thoracic and lumbar levels of the spine has been performed for living tissue by McGill et al. (1993). Using magnetic resonance imaging (MRI) techniques, image slices 5mm thick were generated for 15 young male subjects, although for each subject the level of sectioning was different due to subject variability in height. Slices closest to the disc centre were therefore selected as representative of the muscle at that particular spinal level. Anatomical cross sectional areas were obtained and were corrected to physiologic cross sectional areas by correcting for the angle of fibre orientation relative to the plane perpendicular to the slice. The study was in three dimensions, thereby requiring correction in two planes. Those muscles whose cross sectional areas were not affected by more than 10% when correcting for fibre angle were reported as anatomical cross sectional areas.

Alternatively detailed study of the muscle cross sectional areas has been performed on cadaver specimens. Macintosh and Bogduk performed detailed studies of the lumbar multifidus (1986) and erector spinae (1987, 1991). Measurement of the volume and length of each fascicle allowed the cross sectional area to be calculated (PCSA = volume / length), the results of which were employed in a later model (Bogduk et al., 1992a). Bogduk et al. (1992b) also performed a detailed study of the psoas muscle, again studying each individual fascicle.
The method employed allowed the cross sectional area of the muscle to be calculated without the need for correcting for fibre angle associated with the imaging techniques of McGill et al. (1993). However, the value reported represents the average cross sectional area for the fascicle and does not indicate the maximum force the muscle can generate. Also, typical of cadaver studies, Bogduk et al., 1992b studied cadavers of elderly subjects in excess of 60 years of age. Tissue atrophy is likely to have occurred so that the area of the muscle fascicles measured under-estimates that of a healthy active individual (McGill et al., 1993). However, these cadaver studies provide more data about each individual muscle fascicle than the total cross sectional area of the muscle at each level reported by McGill et al. (1993.)

3.6.3.7.1.2 Cervical Muscles

The focus of study on the lumbar regions has resulted in less study on the muscles in the thoracic and cervical regions. Also, determination of the cross sectional area of the cervical muscles is more difficult due to their comparatively small size. However, studies have been performed due to the requirement for input data for head and neck models. Again these have been performed on living and cadaver tissue.

3.6.3.7.1.2.1 Cadaveric Studies

Within the literature Deng et al. (1987) report the values for the cervical muscle forces applied in their head and neck model. These were estimated using anatomic textbook drawings based upon cadaver studies. However, the values reported represented the average cross section rather than the maximum and did not account for fibre angle. This may overestimate the cross sectional area of oblique muscles.
A review of the muscle cross sectional area in the literature (Yamaguchi et al., 1990) also reported data from two additional studies:

i) Frievalds et al. (1985) determined points of attachment and muscle cross sectional areas for muscles acting at 15 segments of the whole body. Muscles of the trunk were considered relative to the pelvis, centre torso or upper torso. The origin and insertion co-ordinates of the muscles were measured with respect to the co-ordinate frame of the appropriate parts. However, as a secondary information source this makes it difficult to determine the precise levels these measurements correspond to in the co-ordinates frame of this model. Also, it is not certain as to whether these measurements correspond to a single level, a maximum, or an average value.

ii) Values included in the model of the head, neck and upper torso by Daru (1989) were obtained, some from cross sectional measurements and others estimated based upon relative sizes. Pennation angles are given as 10 degrees (for lack of better data). These values corresponded to the C7 vertebral level.

A more detailed study of the spinalis capitis muscle was performed by Martin (1994) through cadaver dissection. Unlike anatomic text books which define this muscle to coincide with semispinalis capitis, Martin (1994) found a distinct muscle running deeper to semispinalis capitis from the spinous process of C6 to the base of the skull. The widest part of this muscle was measured as 2 cm, thereby providing an indication of the maximum cross sectional area.
Richmond et al. (1997) have performed further cadaveric investigations on the cervical muscles. Ten dissected cadavers, three female and seven male were used and cross sectional areas were determined by immersion techniques. The volume of water displaced by the tissue when immersed was divided by the muscle length to obtain the average physiologic cross sectional area. The range and mean values of all the specimens were reported.

3.6.3.7.1.2.2 Studies on Living Subjects

While the values reported from these cadaver studies may underestimate the actual cross sectional areas of living muscles, some in vivo studies have been performed.

Mayoux- Benhamoux et al. (1989) studied computer tomography (CT) scans for 9 female and 7 male untrained normal adults. The image was taken at the C5 level at which the muscles were thought to be well developed and are therefore likely to be representative of maximum or near maximum. A later study (1994) was also performed for longus coli. CT scans for 36 females aged 22 to 30 years were taken at the C5 level and the value reported was 0.56 cm². However, for both studies the cross sectional area measurements were not corrected for fibre angle.

Conley et al. (1995) used MRI techniques to study differences in muscle signal intensity during exercise for living subjects. In contrast, to CT scans, MRI does not expose the subjects to harmful radiation. A total of 10 images were taken covering the length of the neck. Four contiguous images which clearly showed the muscles in each image were selected and the cross sectional area determined. However, the authors were unable to distinguish between each individual muscle group so that values for pairs of muscles were reported. In addition, no correction for fibre pennation was made and the values were averaged rather than reported as a maximum.
3.6.3.7.1.2.3 Additional Study

The variation in techniques used, the difference between living and cadaver tissue and the biological variation between individuals results in a considerable range of data for the cross sectional area of each muscle group. The results of each study are affected by a number of factors:

i) maximum, average or single level value reported
ii) living or cadaver tissue
iii) age of specimen
iv) biological variation between specimens
v) correction for fibre angle
vi) individual fascicles or total muscle considered
vii) individual muscle or combined muscles considered

From this available data it was required to select nominal values which could represent the relative force generating capacity of each muscle. To supplement this a study was performed, comparing relative sizes of each muscle group using scaled anatomy drawings of Eccleshymer and Shoemaker (1911). Muscle groups for which data was not available in the literature were also studied. 2mm graph paper was used to estimate the area of each section of muscle at as many levels for which the muscle was visible. Scaling the drawings to their original size required a scaling factor of 5/4. For some levels both left and right values were observed. The maximum value from these levels (left or right) was then identified.
3.6.3.7.2 Selection of Nominal Cross Sectional Area Values

The overall study of the muscle cross sectional areas indicated muscle groups of similar size and muscle groups which were comparatively larger or smaller. Selection of a nominal cross sectional area value for each muscle group from the data available was guided by these relative proportions.

Data for some muscles showed a large variation. A nominal value representing this range was therefore determined rather than selecting a reported value. For example for semispinalis cervicis a considerable difference between values was found (0.718 - 3.15 cm², mean 1.94 cm²). Values of 1.89 cm² and 2 cm² reported respectively by Daru (1989) and Frievalds (1985) were of closer agreement and also lie closer to the mean. A nominal value of 2 cm² was therefore used.

The selected or estimated nominal values are reported in Table 3-5. As observed in the additional cross sectional anatomy study performed, semispinalis capitis shows greatest cross sectional area in the cervical region. Spinalis capitis, splenius capitis and spinalis cervicis are also of considerable area.

Longissimus capitis, longus capitis, longus coli, longissimus cervicis, splenius cervicis are also of similar area, along with obliquus capitis superior and inferior and rectus capitis posterior major and minor which attach between atlas, axis and the head.

The approximate area of spinalis thoracis was also twice that of semispinalis thoracis, reflecting the same proportions as spinalis cervicis to semispinalis cervicis.
Table 3-5: Selected nominal values for the cross sectional areas of muscles from the cervical studies.

<table>
<thead>
<tr>
<th>Muscle Name</th>
<th>Study</th>
<th>Cross Sectional Area (mm²)</th>
<th>Equivalent Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Longissimus capitis</td>
<td>Approximation</td>
<td>80</td>
<td>80</td>
</tr>
<tr>
<td>Longissimus cervicis</td>
<td>Deng (1989)</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>Splenius cervicis</td>
<td>Deng, (1989)</td>
<td>85</td>
<td>85</td>
</tr>
<tr>
<td>Splenius capitis</td>
<td>Richmond et al. (1997)</td>
<td>426</td>
<td>426</td>
</tr>
<tr>
<td>Spinalis capitis</td>
<td>Martin, (1994)</td>
<td>200</td>
<td>200</td>
</tr>
<tr>
<td>Spinalis cervicis</td>
<td>Own measurements</td>
<td>240</td>
<td>240</td>
</tr>
<tr>
<td>Rectus capitis anterior</td>
<td>Daru, (1989)</td>
<td>13</td>
<td>13</td>
</tr>
<tr>
<td>Rectus capitis lateralis</td>
<td>Daru, (1989)</td>
<td>21</td>
<td>21</td>
</tr>
<tr>
<td>Obliquus capitis superior</td>
<td>Richmond et al. (1997)</td>
<td>103</td>
<td>103</td>
</tr>
<tr>
<td>Obliquus capitis inferior</td>
<td>Richmond et al. (1997)</td>
<td>129</td>
<td>129</td>
</tr>
<tr>
<td>Rectus capitis posterior major</td>
<td>Richmond et al. (1997)</td>
<td>93</td>
<td>93</td>
</tr>
<tr>
<td>Rectus capitis posterior minor</td>
<td>Richmond et al. (1997)</td>
<td>50</td>
<td>50</td>
</tr>
<tr>
<td>Semispinalis cervicis</td>
<td>Approximation</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>Semispinalis capitis</td>
<td>Richmond et al. (1997)</td>
<td>540</td>
<td>540</td>
</tr>
<tr>
<td>Spinalis thoracis</td>
<td>Frievalds et al. (1985)</td>
<td>105</td>
<td>105</td>
</tr>
<tr>
<td>Semispinalis thoracis</td>
<td>Own measurements</td>
<td>58</td>
<td>58</td>
</tr>
<tr>
<td>Longus coli</td>
<td>Own measurements</td>
<td>81</td>
<td>81</td>
</tr>
<tr>
<td>Longus capitis</td>
<td>Richmond et al. (1997)</td>
<td>92</td>
<td>92</td>
</tr>
</tbody>
</table>
3.6.3.7.2.1 Unavailable Data

For some muscles data was not available. For the iliocostalis cervicis muscle, the measurement using the additional cross sectional anatomy study was 1.0 cm$^2$, but this did not account for the significant oblique direction of these fibres. This value may therefore have been an over-estimation. Areas of the thoracic and lumbar fascicles of longissimus thoracis ranged from 0.29-1.78 cm$^2$ and of iliocostalis thoracis from 0.29-1.89 cm$^2$. Assuming a similar distribution between longissimus thoracis and longissimus cervicis and iliocostalis thoracis and iliocostalis cervicis, the fascicles of iliocostalis cervicis and longissimus cervicis may be of similar areas. The iliocostalis cervicis muscle was therefore assigned a nominal value of 0.75cm$^2$.

3.6.3.7.2.2 Intersegmental Muscles

Data regarding the cross sectional area of the intersegmental muscles which attach between numerous levels of the spine was limiting. For rotatores no data could be found, whereas for the interspinalis and intertransverse single values between the centre and upper torso were reported to be 0.5 and 0.25 cm$^2$ respectively (Frievalds, 1985). These single values do not represent any variation from level to level which is likely to occur for these muscles. Also, the values reported for the cervical and thoracic multifidus ranged from 1.25 cm$^2$ (Frievalds, 1985), 0.35 cm$^2$ (own study) to 5.80 cm$^2$ (Mayoux Benhamou et al., 1994). However, the additional study using cross sectional anatomy drawings did not identify a large area for the multifidus muscle at the C5 level. The large value reported by Mayoux Benhamou et al. (1994) may possibly have included an additional muscle not distinguishable by the CT scans.
Due to the deep nature of these muscles lying close to the spine, it was assumed these muscles were relatively small. Nominal values for the intersegmental muscles were taken at each level of attachment to be:

- interspinalis 0.50 cm\(^2\)
- intertransverse 0.25 cm\(^2\)
- cervical and thoracic multifidus 0.30 cm\(^2\)
- rotatores 0.30 cm\(^2\)

Variations in magnitude and distribution were investigated in a separate sensitivity analysis study.

3.6.3.7.2.3 Lumbar Muscle Forces

For the lumbar muscle forces, the data reported by Bogduk et al. (1992a, 1992b) was selected on account that it provided information regarding each individual muscle fascicle. This was appealing for the derivation of the distributed force pattern in the model. Alternatively quadratus lumborum which was not included in the detailed studies was represented using data from McGill et al. (1993).
3.6.3.7.2.4 Abdominal Muscles

For the abdominal muscles, the study by Dumas et al. (1991) from which the points of attachment and lines of action were estimated did not report the corresponding cross sectional areas. Alternatively McGill et al. (1993) estimated the cross sectional area of these muscles at the lumbar levels using MRI and corrected for fibre pennation. The maximum values reported were 1121mm$^2$ for external oblique, 1154mm$^2$ for internal oblique, 802mm$^2$ for rectus abdominis and 646mm$^2$ for transverse abdominis. These total cross sectional areas were therefore used in this study.

3.6.3.7.3 Force Magnitude

The data presented represents the measured cross sectional areas for muscles on right or left sides of the body, or the average. Assuming sagittal symmetry, the total muscle area in this plane is therefore twice the nominal value. Also, assuming a force coefficient of 50N/cm$^2$ (this lies within the range of coefficients reported in the literature), the total force generated by each muscle group (or fascicle) is $2 \times 50 \text{ (N/cm}^2\text{)} \times \text{cross sectional area (cm}^2\text{)}$, which is equivalent to the nominal cross sectional area values in this case. Hence, for each muscle group, selection of the nominal maximum cross sectional allows an estimate of the maximum force generating capacity of each muscle to be obtained. The relative contribution of each individual muscle group towards the stability of the spine can therefore be investigated quantitatively using these nominal values.
For this study, nominal values selected are based upon the mean values reported by the authors of the individual studies. However, considerable biological variation occurs. For example in the study of McGill et al. (1993) the total cross section for the abdominal wall is reported as 1104mm at the L5 level with a standard deviation of 393mm. On account of this the single nominal values selected were not intended to represent the large biological range associated with the muscle tissue of each individual. Rather the nominal values were to reflect the relative strengths of the muscle groups. No attempt was therefore made to increase the cross sectional area values reported in cadaver studies to account for greater cross sectional area in living tissue.

3.6.3.7.4 Force Distribution

Whereas the individual fascicles of the lumbar muscles had been measured separately, the measurements for many muscle groups represented the muscle as a whole. The distribution of these forces was therefore not known. A uniform distribution was therefore initially assumed. Thus, for the longus capitis muscle, the equivalent force of 92N was divided among the four fascicles so that each generated a force of 23N. The effects of force distribution are considered in a later sensitivity analysis.
3.6.4 LIGAMENT FORCES IN THE SAGITTAL PLANE

During flexion or extension, the points of attachment on the vertebrae for the ligaments move apart. The ligaments can therefore be stretched and a tensile force generated within them. Thus, unlike the muscle elements which must contract and expend energy to produce a force, the ligaments can be stretched passively and generate a tensile force. This force pulls the vertebrae together and could contribute to the compressive thrust required for the stability of the spine when represented as an arch structure.

However, Aspden (1992) has proposed the ligaments cannot exert a tensile force in the sagittal plane unless their collagen fibres are aligned when stretched in this direction. Based on a morphologic study (Hukins et al., 1990), only the anterior and posterior longitudinal and the ligamentum flavum ligaments appear capable of generating significant force. Of these the anterior ligament requires extension of the vertebrae in order to do so.

The tensile force generated within the ligament can be estimated based on the relation:

\[ \text{force} = \text{stress} \times \text{cross sectional area}. \]

Values for the cross sectional area of the ligaments and their corresponding stress when stretched in a given posture are needed. Stress can be estimated using experimentally derived stress strain-curves.
3.6.4.1 Estimation of Ligament Cross Sectional Area

Chazal et al. (1985) investigated the cross sectional areas of human spinal ligaments obtained from fresh cadaver specimens (less than 24 hours after death), ranging from 30 to 80 years in age, with an average of 53 years. The cross section was measured every 2mm along the entire length of the ligament with a palpator. These values were reported for levels ranging from C2 to L4 for the posterior longitudinal ligament and from T4 to L4 for the ligamentum flavum. However, values were not reported at every level in between.

Panjabi et al. (1991c) also performed a detailed study for the lumbar spine ligaments, using fresh cadaver specimens, although the age details of these specimens were not provided. They reported a number of fibres existed for each ligament, each with distinct points of attachment. The co-ordinates for each pair of fibre attachment points, the cross sectional area, direction of the fibre and fibre length were determined. The cross sectional areas were determined by digitising photographs taken for the various ligament fibre cross sections. The average values for these measurements for each ligament were then determined.

The variation in values reported by these two studies and the lack of data at several levels makes it difficult to assign a particular value for the cross sectional area of the ligaments. The values selected are shown in Table 3-6. Values were estimated (based on interpolation) for the levels at which measurements had not been obtained, assuming cross sectional area increases descending the spine.
To use this data to derive a distributed force pattern for these ligaments, the data reported by Chazal et al. (1985) for the cervical and thoracic levels was used, along with the complete set of data reported by Panjabi et al. (1991c) for the lumbar levels.

Table 3-6: Cross sectional areas for the posterior longitudinal ligament and ligamentum flavum.

<table>
<thead>
<tr>
<th>Vertebral level</th>
<th>Cross sectional areas for ligamentum flavum (mm$^2$), $\theta$=estimated.</th>
<th>Cross sectional areas for posterior longitudinal ligament (mm$^2$), $\theta$=estimated.</th>
</tr>
</thead>
<tbody>
<tr>
<td>C2/C3</td>
<td>(20)</td>
<td>10</td>
</tr>
<tr>
<td>C3/C4</td>
<td>(20)</td>
<td>(10)</td>
</tr>
<tr>
<td>C4/C5</td>
<td>(20)</td>
<td>(10)</td>
</tr>
<tr>
<td>C5/C6</td>
<td>(20)</td>
<td>(10)</td>
</tr>
<tr>
<td>C6/C7</td>
<td>(20)</td>
<td>(10)</td>
</tr>
<tr>
<td>C7/T1</td>
<td>(20)</td>
<td>11</td>
</tr>
<tr>
<td>T1/T2</td>
<td>(20)</td>
<td>(11)</td>
</tr>
<tr>
<td>T2/T3</td>
<td>(20)</td>
<td>(14)</td>
</tr>
<tr>
<td>T3/T4</td>
<td>(20)</td>
<td>(14)</td>
</tr>
<tr>
<td>T4/T5</td>
<td>34</td>
<td>17</td>
</tr>
<tr>
<td>T5/T6</td>
<td>26</td>
<td>(17)</td>
</tr>
<tr>
<td>T6/T7</td>
<td>29</td>
<td>(17)</td>
</tr>
<tr>
<td>T7/T8</td>
<td>25</td>
<td>(17)</td>
</tr>
<tr>
<td>T8/T9</td>
<td>19</td>
<td>(19)</td>
</tr>
<tr>
<td>T9/T10</td>
<td>(25)</td>
<td>(20)</td>
</tr>
<tr>
<td>T10/T11</td>
<td>30</td>
<td>(20)</td>
</tr>
<tr>
<td>T11/T12</td>
<td>(40)</td>
<td>(20)</td>
</tr>
<tr>
<td>T12/L1</td>
<td>(40)</td>
<td>24.2</td>
</tr>
<tr>
<td>L1/L2</td>
<td>56.7</td>
<td>36.5</td>
</tr>
<tr>
<td>L2/L3</td>
<td>73.5</td>
<td>20.8</td>
</tr>
<tr>
<td>L3/L4</td>
<td>71.7</td>
<td>22.2</td>
</tr>
<tr>
<td>L4/L5</td>
<td>78.7</td>
<td>51.8</td>
</tr>
</tbody>
</table>

3.6.4.2 Estimation of Ligament Stress

Chazal et al. (1985) tested the response of the ligaments under loads applied at a constant slow rate of 1mm/min. The resulting load deformation curves had a sigmoid shape with 3 distinct portions: OA, AB and BC shown in Figure 3-45. The stress-strain values at the most characteristic points (A, B and C) were reported.
Based upon the strain in the ligaments due to flexion, the resulting stress can therefore be estimated and the consequent force for a given cross section can be determined.

Figure 3-45: Three distinct portions of a load deformation curve for the ligaments under loading applied at a slow rate (1mm / min) reported by Chazal et al. (1985).
To estimate the ligament strains for each flexed posture in the model, the distance between their points of attachment was measured. The ligamentum flavum was assumed to be attached between the lamina of adjacent vertebrae. The points of attachment were represented on the anterior vertebral body surface approximately at 4/5 vertebral height on the inferior vertebra and 1/5 vertebral height on the superior vertebra. The posterior longitudinal ligament points of attachment were represented at the posterior vertebral body surface level with the inferior surface of the vertebral body above and the posterior superior surface of the vertebral body below. These representative points of attachment were similar to those used in previous finite element models (e.g. Shirazi Adl et al., 1987, see page 109, Figure 2-25).

The change in distance between these points due to flexion was assumed to represent the change in ligament length. Expressing this as a percentage of the length values for the erect posture represented the resulting strain due to flexion.

However, due to lack of data in the literature regarding precise vertebral configurations, the relative rotations of the vertebrae had been estimated to approximate flexed postures. Estimation of the ligament lengths for these postures is not supported by a sufficient level of anatomic detail. Consequently, unrealistic lengths and strain values were found. An alternative method of study for the ligament forces was therefore required.
3.6.4.4 Application of theoretical Ligament Forces to the Model Developed in ADAMS

The amount of force the ligaments are theoretically capable of generating at each of the points on the characteristic stress-strain curve (Figure 3-45) was determined. The resulting estimated forces for the posterior longitudinal ligament and the ligamentum flavum are shown in Table 3-7. By assuming a certain degree of strain for a given posture, the corresponding forces can be obtained. Ligament strains were considered for points A and B in the erect and flex1 postures and for points B and C in the flex2 and flex3 postures.

Table 3-7: Estimated forces for the ligamentum flavum and posterior longitudinal ligament.

<table>
<thead>
<tr>
<th>Vertebral level</th>
<th>Force (N) Ligamentum Flavum</th>
<th>Force (N) Posterior Longitudinal Ligament</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>At point A</td>
<td>At point B</td>
</tr>
<tr>
<td>C2/C3</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>C3/C4</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>C4/C5</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>C5/C6</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>C6/C7</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>C7/T1</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>T1/T2</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>T2/T3</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>T3/T4</td>
<td>63.4</td>
<td>271.4</td>
</tr>
<tr>
<td>T4/T5</td>
<td>107.78</td>
<td>461.38</td>
</tr>
<tr>
<td>T5/T6</td>
<td>82.42</td>
<td>352.82</td>
</tr>
<tr>
<td>T6/T7</td>
<td>91.93</td>
<td>393.53</td>
</tr>
<tr>
<td>T7/T8</td>
<td>79.25</td>
<td>339.25</td>
</tr>
<tr>
<td>T8/T9</td>
<td>60.23</td>
<td>257.83</td>
</tr>
<tr>
<td>T9/T10</td>
<td>79.25</td>
<td>339.25</td>
</tr>
<tr>
<td>T10/T11</td>
<td>95.1</td>
<td>407.1</td>
</tr>
<tr>
<td>T11/T12</td>
<td>126.8</td>
<td>542.8</td>
</tr>
<tr>
<td>T12/L1</td>
<td>126.8</td>
<td>542.8</td>
</tr>
<tr>
<td>L1/L2</td>
<td>179.74</td>
<td>769.42</td>
</tr>
<tr>
<td>L2/L3</td>
<td>232.10</td>
<td>997.40</td>
</tr>
<tr>
<td>L3/L4</td>
<td>227.29</td>
<td>972.97</td>
</tr>
<tr>
<td>L4/L5</td>
<td>249.48</td>
<td>1067.96</td>
</tr>
</tbody>
</table>
3.6.4.5 Alternative Points of Attachment

The posterior longitudinal ligament was assumed to be attached between pairs of adjacent vertebrae, in accordance with anatomic textbook descriptions and more detailed finite element studies (e.g. Shirazi Adl et al., 1986). However, Panjabi et al. (1991c) studied ligaments of the lumbar region and reported the posterior longitudinal ligament to have three layers of fibres ranging from superficial which spanned two to three layers, to the deep layer with intersegmental attachments. Data regarding the cross sectional areas for each layer was however not reported. Consequently, in this study only the deeper layer was considered.

To consider the effects of a multi-layered ligament, the tests were repeated using points of attachment in the lumbar regions spanning two vertebral levels.
3.6.5 FORCES DUE TO INTRA ABDOMINAL PRESSURE

Aspden (1987) described intra abdominal pressure acting over the anterior surfaces of the lumbar vertebrae and discs. The direction of each force was perpendicular to the axes of the corresponding vertebra or disc at that level. The force values applied had been determined in a previous study reported in the literature based upon experimental techniques. In the upright posture for subjects performing Valsalva’s manouvre the value for each vertebral force was 10N and each disc force 4N. This was equivalent to an intra abdominal pressure of 4KPa. These values were increased by a factor of 2.5 for a flexed posture with trunk angle 30 degrees, and by a factor of 16 in the flexed posture when carrying a 900N load.

Also, Schultz et al. (1982) measured intra abdominal pressure for living subjects and reported an intra abdominal pressure of 1KPa in relaxed standing, 4.2 KPa with the trunk flexed at 30 degrees when holding the arms out and 4.4 KPa with the trunk flexed, arms out and holding an 8Kg load.

A range of intra abdominal pressure were therefore investigated. Values for the 30 degree trunk flexion posture were assumed to correspond to the flex2 posture in the model investigations. Based upon interpolation the maximum intra abdominal pressure in the flex3 posture would be 10KPa and 18 KPa when holding a load. The final range of pressures are shown in Table 3-8.

Table 3-8: Nominal values of intra abdominal pressure applied.

<table>
<thead>
<tr>
<th>Body weight only</th>
<th>Additional 900N load</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect 1KPa and 4KPa</td>
<td>12KPa</td>
</tr>
<tr>
<td>Flex1 6KPa</td>
<td>14KPa</td>
</tr>
<tr>
<td>Flex2 8KPa</td>
<td>16KPa</td>
</tr>
<tr>
<td>Flex3 10KPa</td>
<td>18KPa</td>
</tr>
</tbody>
</table>
4. INVESTIGATING STABILITY AND LOADING OF THE SPINE USING THE DEVELOPED MODEL

In Chapter three, the representation of the spine as an arch, but with forces distributed along the entire length of the spine was discussed. The weight applied by the head and neck muscle forces provides the necessary compressive thrust at this end for the spine to be considered as an arch in both the erect and flexed postures.

Using the derived loading patterns, body weight, muscle and ligament forces could be applied along the curvature of the whole spine. The parallel component of muscle and body weight forces contributes to the compressive thrust required. However, perpendicular components also affect the thrustline curve. To determine the effects of these forces on the stability of the spine, a series of investigations were performed. For each study, stability of the spine was assessed by comparison of the path of the predicted thrustline with the path of the vertebral body anterior and posterior borders, and their centre lines.

i) Stability of the Spine Under the Influence of Body Weight
An initial investigation was performed to determine the effects of body weight on the stability of the spine as an arch. An indication of the amount of muscle and ligament force required for support, and the region of application of this force was then obtained.
In addition to the applied body weight forces, the effects of a 900N load lifted in the arms on the stability of the spine were investigated. Forces of magnitude 300N acting at each of the T2, T3 and T4 levels were applied to the model.

Using the nominal maximal muscle forces defined for each muscle group, the role of each individual muscle group in contributing towards the stability of the spine when supporting body weight was considered. A comparison of the effects of each muscle group allowed the relative importance of each in its contribution to stability to be determined.

iv) The Contribution of the Ligaments Towards the Stability of the Spine
The role of the ligaments in supporting the spine during lifting activities were investigated. A comparison between the potential role of the ligaments and muscles in contributing to the stability of the spine was then made.

v) Sensitivity Analysis: Variation in Muscle Force
Due to lack of data in the literature regarding the properties of spinal muscles, some assumptions had been made for the magnitude, direction and distribution of force. The effects of these assumptions were therefore investigated in separate studies described below.

a) The Effects of Spinal Curvature on Muscle Force Direction
A comparison of the representation of muscle lines of action as a straight line between their points of attachment, or along a deviated path accounting for the spinal curve between the muscle points was made. Due to the number of muscle groups tested, this study was limited to those which spanned several vertebral levels and were most likely to be affected by curvature.
b) The Effects of Muscle Force Distribution on the Predicted Thrustline
Lack of data in the literature regarding the force generating capacity of each individual muscle fascicle was approached by assuming a uniform distribution of force. To test for the effects of variation in force distribution, patterns of increasing or decreasing force descending the spine, or with the greatest maximum area in fascicles which span the middle region were also investigated. Representative muscle groups were semispinalis capitis (cervical region), spinalis thoracis (thoracic region) and intersegmental muscles interspinalis, rotatores and cervical and thoracic multifidus.

c) The Effects of Muscle Force Magnitude on the Predicted Thrustline
Sparsity of literature data for the intersegmental muscles required the force generating capacity to be estimated. To investigate the possibility of greater force generating capacity, the effects of increasing the magnitude of these muscle forces were therefore considered.

vi) Combinations of Muscle Forces which Satisfy the Stability of the Spine when Supporting Body Weight, and an External Load of 900N
Based on the predicted behaviour for each individual muscle group, combinations of muscle forces which satisfied stability of the spine for each posture were sought. The results of this study provided an indication of the ability of the spinal muscle forces to stabilise the spine in a given posture, when supporting body weight, and a 900N external load.
4.1 THE ACTION OF BODY WEIGHT ON THE SPINE REPRESENTED AS AN ARCH STRUCTURE

For each posture, the predicted thrustline due to the applied distributed loading pattern for body weight was determined. The outlines of the vertebral body region relative to the reference line, and the predicted thrustline were plotted to provide a graphical indication of the state of stability for the spine.

4.1.1 MODEL PREDICTIONS

For all postures, the consecutive portions of the thrustline followed an overall path which approximated a posteriorly convex curve, and did not coincide with the path of the vertebral bodies at any level. Instability was therefore predicted for all postures.

Figure 4-1 shows the predicted thrustline due to body weight and the vertebral body outlines for the erect posture. The single thrustline curve did not match the posterior concave curves of the vertebral bodies in the cervical and lumbar regions, while the depth of the thrustline curve was greater than that of the posterior convex thoracic vertebral curve.

Similarly for each of the flexed postures in which the spine tended towards a single posterior convex curve, the depth of the thrustline curve was greater than the curve followed by the vertebral bodies. This difference increased with flexion. Figure 4-2 shows the predicted thrustline for the flex2 posture illustrating the greater depth of curvature.
Figure 4-1: Predicted thrustline due to body weight and the outlines of the vertebral limits relative to the reference line for the erect posture.

Figure 4-2: Predicted thrustline due to body weight and the outlines of the vertebral limits relative to the reference line for the flex2 posture.
4.1.2 EXPLANATION OF THE PREDICTED THRUSTLINES

4.1.2.1 Force Direction

The forces due to body weight act vertically downwards due to gravity, regardless of spinal configuration. The distributed loading pattern therefore represents a set of uni-directional forces. However, the direction of these forces changes relative to the vertebral axes and the reference line according to posture. For all postures including the erect, the reference line shows a forward inclination by varying degrees. The force components are therefore negative in both the perpendicular and parallel directions relative to A. This is illustrated in Figure 4-3.

![Figure 4-3: The negative components of force due to body weight perpendicular and parallel to a forward inclined reference line AB.](image)

Figures 4-4 to 4-7 illustrate the direction of body weight forces relative to the reference line for each posture. With flexion, the inclination of the reference line is increased, and consequently, the perpendicular components of body weight force are increased, and the parallel components proportionally decreased. The end support reaction component in the parallel direction (H1) is therefore also decreased.
Figure 4-4: The points of application and line of action of forces due to body weight along the spine relative to the reference line in the erect posture.
Figure 4-5: The points of application and line of action of forces due to body weight along the spine relative to the reference line in the flex1 posture.

Figure 4-6: The points of application and line of action of forces due to body weight along the spine relative to the reference line in the flex2 posture.
4.1.2.2 Changes in Point of Force Application with Posture

In the erect posture, the forces due to the weight of the head and the weight of the trunk in the thoracic and lumbar regions act anterior to the reference line, while in the cervical and upper thoracic regions the weight is posterior. The head weight also acts superior to the end supports A and B. These points of force application are similar for the flex1 posture.

However, in the flex2 and flex3 postures, the increased curvature of the spine moves the forces due to body weight posterior relative to the reference line. This affects the moment arm of the parallel component of force, and alters the sign of the moments produced by these components about A and B. However, the decrease in this force component with flexion reduces the moment generated so that this effect is small.
Also, with flexion, the increase in curvature results in a slight reduction of the length of the reference line AB. Moments generated about the ends A and B by the perpendicular components of force are therefore reduced slightly, resulting in a slight reduction in R1 and R2. However, the increase in this force component with flexion results in an overall greater moment generated so that the effects of the reduction in moment arm are again small.

4.1.2.3 Generated Moments

4.1.2.3.1 Erect Posture

In the erect posture, the dominant components of force were in the parallel direction and the perpendicular components of force were in proportion [0.02:1]. Despite a variation in force magnitude and moment arm at each vertebral level, the dominant moments were generated by the large parallel components of force. Indeed the actual values calculated for the moments generated due to perpendicular components were 2368.51 Nmm about A and -2034 Nmm about B, whereas those generated about A or B due to parallel components were 9698.85 Nmm.

Due to regional variation in point of force application anterior, posterior (and the head weight superior) to the reference line, positive and negative moments are generated. Table 4-1 shows the signs (positive or negative) for the moments generated about the ends of the spine by the forces applied in the various regions.
Table 4-1: The calculated signs of the moments generated by body weight and the consequent end support reaction components R1 and R2.

<table>
<thead>
<tr>
<th></th>
<th>Parallel Component About A</th>
<th>Parallel Component About B</th>
<th>Perpendicular Component About A</th>
<th>Perpendicular Component About B</th>
<th>R1</th>
<th>R2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>positive</td>
<td>positive</td>
<td>positive</td>
<td>positive</td>
<td>negative</td>
<td>positive</td>
</tr>
<tr>
<td>Cervical levels</td>
<td>negative</td>
<td>negative</td>
<td>positive</td>
<td>negative</td>
<td>positive</td>
<td>negative and positive</td>
</tr>
<tr>
<td>Thoracic levels</td>
<td>positive</td>
<td>positive</td>
<td>positive</td>
<td>negative</td>
<td>negative and positive</td>
<td>positive</td>
</tr>
<tr>
<td>Lumbar levels</td>
<td>positive</td>
<td>positive</td>
<td>positive</td>
<td>negative</td>
<td>negative and positive</td>
<td>positive</td>
</tr>
</tbody>
</table>

The weight of the head and the trunk at lower thoracic and lumbar levels is much greater than that at cervical levels and has a greater moment contribution. Also, the combined weight at the thoracic and lumbar levels has a greater influence on the total moment than that generated by the head acting superior to B. The dominant moments are therefore generated by the parallel components of force anterior to the reference line in the thoracic and lumbar regions. Using the relations:

\[ R1 = -(\text{sum of total moment about B}) \]
\[ R2 = (\text{sum of total moment about A}) \]

this makes R1 negative and R2 positive. The predicted values for the end support reaction components are reported in Table 4-2.

Table 4-2: Calculated values for the end support reaction components (N) due to body weight forces.

<table>
<thead>
<tr>
<th>Posture</th>
<th>R1</th>
<th>R2</th>
<th>H1</th>
<th>H2</th>
<th>R1/H1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect</td>
<td>-14.32</td>
<td>22.54</td>
<td>379.87</td>
<td>-34.59</td>
<td>0.0377</td>
</tr>
<tr>
<td>Flex1</td>
<td>62.32</td>
<td>84.24</td>
<td>350.56</td>
<td>-31.92</td>
<td>0.1778</td>
</tr>
<tr>
<td>Flex2</td>
<td>142.17</td>
<td>146.81</td>
<td>246.70</td>
<td>-22.46</td>
<td>0.5763</td>
</tr>
<tr>
<td>Flex3</td>
<td>170.85</td>
<td>195.67</td>
<td>100.14</td>
<td>-9.12</td>
<td>1.7061</td>
</tr>
</tbody>
</table>
4.1.2.3.2 Flexed Postures

With flexion, due to the increase in perpendicular components of force, the dominant moments are generated by these components. Indeed for the flex1 posture the total moment generated by the perpendicular components is $-3.56 \times 10^4$ Nmm about A and $4.27 \times 10^4$ Nmm about B, whereas the total moment generated by the parallel component about A or B is 2282.62 Nmm.

The combined weight at all vertebral levels has a greater influence on the total moment than that generated by the head acting superior to B. Using Table 4-1, the total moment generated by both parallel and perpendicular components of force is positive about A, and negative about B. This makes both R1 and R2 positive.

Similarly for the flex2 and flex3 postures, R1 and R2 (reported in Table 4-2) were both positive. However, due to the increase in perpendicular components, these values are greater to ensure equilibrium. The proportional decrease in parallel components of the end support reactions (H1 and H2) with flexion can also be observed.
4.1.2.4 Effects of Forces and Moments on Predicted Thrustline

4.1.2.4.1 Erect Posture

In the erect posture, the negative value of $R_1$ determined the projection of the initial portion of the thrustline in the direction of the negative $x$ co-ordinate axis.

At each point of force application ($i$), the perpendicular ($F_{ri}$) and parallel ($F_{hi}$) components are added to the perpendicular and parallel components of the compression force vector. At the point of force application ($k$) the direction of the resultant force vector ($\theta_i$) is determined by the expression

$$\tan\left(\frac{R_1 + \sum_{i=1}^{k} F_{ri}}{H_1 + \sum_{i=1}^{k} F_{hi}}\right).$$

The addition of a negative perpendicular component to a negative value of $R_1$ makes the nominator increasingly negative, while the addition of a negative parallel component to the large positive value of $H_1$ (representing the total force in this direction) decreases the denominator but does change it from being positive. The slope of the thrustline at each stage increases in the direction of the negative $x$ co-ordinate axis (Figure 4-8).
Figure 4-8: Local application of force due to body weight alters the direction of the thrustline in the erect posture.

Since the forces due to body weight act vertically downwards and are therefore inclined by the same amount relative to A, the proportion [Fr:Fh] does not change. There is thus a continuous increase in the thrustline slope, creating a curved path. Shearing of this line relative to AB however transfers the curve to the positive x co-ordinate axes (Figure 4-9).

Figure 4-9: A thrustline with increasing negative gradient lies on the positive x co-ordinate axis when sheared relative to AB.
4.1.2.4.2 Flexed Postures

In the flex1 posture, due to the positive value of R1 the initial projection of the thrustline is posterior to the reference line.

The initial proportion $R1/H1$ is [0.1778:1] (Table 4-2). The addition of body weight forces with negative perpendicular and parallel components in proportion [0.4:1] results in a greater reduction in the perpendicular component which eventually becomes negative. Consequently, the slope of the thrustline at each stage negatively increases, resulting in a posterior convex curve (Figure 4-10). A shear transformation maintains this curve posterior to the reference line. Also, the larger proportion of the perpendicular component in this ratio means a greater change in slope between consecutive thrustline portions than was obtained for the erect posture. Consequently, the curvature of the thrustline is steeper in the flex1 posture due to an increase in the perpendicular component.

![Diagram](image)

Figure 4-10: Local application of force due to body weight alters the direction of the thrustline in the flexed postures.
Due to positive values for R1 and R2 for the flex2 and flex3 postures, the initial thrustline projection is also on the positive x co-ordinate axis. However, the proportion of perpendicular to parallel components increases to [1.17:1] in the flex2 posture, and to [3.66:1] in the flex3 posture. This increases the initial projection of the thrustline in the perpendicular direction (Table 4-2 shows significantly larger values for R1 and R2), but also results in a greater change in thrustline direction at each point of application. Consequently, the curvature of the thrustline is increased.
4.2 THE ACTION OF EXTERNAL LOADS

Forces of 300N each were applied at T2, T3 and T4 levels to represent a 900N load lifted in the arms, in addition to the existing distributed body weight pattern. The effects of these forces combined with the existing body weight force on spinal stability were then determined.

4.2.1 MODEL PREDICTIONS

As for the body weight investigations, the predicted thrustlines for each posture followed a posterior convex curve which did not correspond to the path of the vertebral bodies. Instability was therefore predicted at all vertebral levels for each of the four postures. Also, these curves were much steeper than those predicted for the action of body weight only. Figures 4-11 and 4-12 compare the predicted thrustlines due to body weight with and without external loading for the erect and flex3 postures respectively. Greater instability was predicted with the addition of the external load.
Figure 4-11: Predicted thrustline due to body weight only, and body weight and external load, and the outlines of the vertebral limits relative to the reference line for the erect posture.

Figure 4-12: Predicted thrustline due to body weight only, and body weight and external load, and the outlines of the vertebral limits relative to the reference line for the flex3 posture.
4.2.2 EXPLANATION OF THE EFFECTS OF EXTERNAL LOADS

4.2.2.1 Force Direction

The external load forces due to gravity act vertically downwards regardless of posture and have the same direction as the forces due to body weight. The proportions of the perpendicular and parallel components for these forces are the same as for the body weight forces in each of the four postures.

4.2.2.2 Point of Force Application

In the erect posture, the points of application of the external load at the centre of the T2, T3 and T4 vertebrae are slightly posterior to the reference line. The moments generated by the parallel components of the external load applied at these points are therefore negative about A and B.

With flexion, the curvature of the spine is increased and the centres of the T2, T3 and T4 vertebral bodies are moved further from the reference line of the spine. The moment arm of the parallel components of force due to the external load is therefore increased and the moment generating capacity is increased. Table 4-3 shows the moment arms for these parallel components of force in each posture. Due to the magnitude of these forces and the consequent moments, this is more significant than for the forces due to body weight.
Table 4-3: The eccentricity (mm) of the parallel component of force due to external load. (N.B. positive values indicates posterior positioning relative to the reference line.)

<table>
<thead>
<tr>
<th>Point of Force Application</th>
<th>Eccentricity of parallel component from A or B in erect posture</th>
<th>Eccentricity of parallel component from A or B in flex1 posture</th>
<th>Eccentricity of parallel component from A or B in flex2 posture</th>
<th>Eccentricity of parallel component from A or B in flex3 posture</th>
</tr>
</thead>
<tbody>
<tr>
<td>T2 centre</td>
<td>2.20</td>
<td>28.86</td>
<td>68.25</td>
<td>48.83</td>
</tr>
<tr>
<td>T3 centre</td>
<td>6.43</td>
<td>33.56</td>
<td>75.51</td>
<td>58.31</td>
</tr>
<tr>
<td>T4 centre</td>
<td>9.38</td>
<td>37.67</td>
<td>82.27</td>
<td>67.89</td>
</tr>
</tbody>
</table>

4.2.2.3 Generated Moments

4.2.2.3.1 Erect Posture

In comparison with the body weight forces applied eccentric to the spine, the external load is applied at the centre of T2, T3 and T4 vertebrae, very close to the reference line in the erect posture. The moment arm of the parallel component of force is therefore relatively small (Table 4-3) and consequently, the moments generated are relatively small. Also, these moments oppose the direction of those due to body weight in the thoracic and lumbar regions, and the overall moment due to parallel force components is small.
The perpendicular component of force due to the external load is smaller than the parallel component, although the overall magnitude of force is larger than that due to body weight. The moment arm of the perpendicular component of the force due to the external load is larger than that of the parallel component. As a result of the large moment arm and the greater force magnitude, overall the dominant moment is generated by the perpendicular component of the external load. This is positive about A, and negative about B, resulting in both positive values for R1 and R2. These values are shown in Table 4-4. The application of the external load closer to B results in greater end support reaction at this end, explaining larger values for R2 than for R1.

Table 4-4: End support reaction components (N) calculated due to body weight and 900N external loading.

<table>
<thead>
<tr>
<th>Posture</th>
<th>R1</th>
<th>R2</th>
<th>H1</th>
<th>H2</th>
<th>R1/H1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect</td>
<td>1.36</td>
<td>26.23</td>
<td>1279.66</td>
<td>-34.59</td>
<td>0.001</td>
</tr>
<tr>
<td>Flex1</td>
<td>212.67</td>
<td>281.05</td>
<td>350.56</td>
<td>-31.92</td>
<td>0.607</td>
</tr>
<tr>
<td>Flex2</td>
<td>411.46</td>
<td>562.01</td>
<td>831.05</td>
<td>-22.46</td>
<td>0.495</td>
</tr>
<tr>
<td>Flex3</td>
<td>439.40</td>
<td>795.31</td>
<td>337.34</td>
<td>-9.12</td>
<td>1.303</td>
</tr>
</tbody>
</table>

4.2.2.3.2 Flexed Postures

In the flexed postures, the inclination of the external load and body weight force vectors relative to the reference line increases so that the perpendicular component of force becomes increasingly dominant. The moments they generate therefore increase proportionally and are also dominant.
4.2.2.4 Effects of Forces and Moments on the Predicted Thrustline

For all postures, due to the positive value predicted for R1, the initial portion of the thrustline is projected on the positive x co-ordinate axis relative to the reference line.

For the erect posture due to the very small positive value of R1 (1.36 N), the addition of subsequent negative perpendicular components of force to the compression force vector makes the perpendicular component negative. The parallel component which is initially the sum of the parallel force components (1279.6 N) remains positive but decreases. The slope of each portion of the thrustline is therefore positively decreased, and a curved path is followed.

In the flexed postures, due to the dominant moments generated by the perpendicular components of external load forces, values for R1 and R2 are larger (Table 4-4). Also, due to the magnitude of the external load forces, the change in slope is greatest at these points of force application, resulting in the peak of the thrustline curve at T2, T3 and T4 levels at which the external load is applied. The thrustline is therefore steeper than that due to body weight, and increases with flexion. This is shown in Figure 4-12.
4.2.3 DISCUSSION

For each posture, the spine was predicted to be unstable when supporting the weight of the upper body, and when supporting an additional external load of 900N. Additional muscle force is therefore required to ensure stability.

For each posture, the moments generated by the body weight force components directly affected the end support reactions components R1 and R2. The moments generated by the parallel components of force due to body weight acting anterior to the reference line are positive about the ends A and B, whereas the moments generated by the perpendicular components of force are positive about A and negative about B. The moments about B are therefore reduced due to terms of opposing direction, and consequently, R1 is less than R2.

Muscles acting posterior to the spine generate moments due to parallel components of force which are negative about A and B. The effect of muscle forces will therefore act to reduce R2 and increase R1. This is beneficial to reduce the force generated at the upper cervical level and increase the force at the lumbo-sacral level where greater loads can be withstood.

With the addition of an external load at the upper thoracic levels, the reaction force at the upper end of the spine (R2) was larger than R1. Again, muscle forces will act to alter the proportions of R1 and R2, reducing the loads on the cervical vertebrae.
With flexion, the perpendicular components of force increased and the parallel components decreased. This was accompanied by an increase in both R1 and R2, and a decrease in H1 and H2 for equilibrium. Due to this increase in the perpendicular component of force, a steeper thrustline curve was projected. These effects were increased with the application of an external load. Muscle forces are required to oppose the effects of body weight by generating force with perpendicular components in the opposite direction, or by generating larger parallel components to decrease the proportion of the perpendicular component of the resultant force vectors.

For all cases, the overall thrustline was projected to the right of the reference line as a single posterior convex curve, due to the body weight and external load forces acting in the same direction at each level. Stability of the spine in the erect posture with three distinct alternating curves can therefore not be obtained under the action of body weight and when lifting an additional load. Also, in the flexed postures, although the configuration of the spine tended towards that of a single posterior convex curve, the thrustline curvature was much greater resulting in instability at all levels. Additional forces due to muscle and ligaments are therefore required to flatten the overall thrustline curve, and due to a steeper curve greater muscle and ligament forces are required with flexion and external loading. More specifically, the multi-directional muscle and ligament forces may provide greater control of the direction of the thrustline path so that it corresponds to the configuration of the vertebral bodies, particularly when cervical and lumbar lordosis are present.
4.3 THE EFFECTS OF THE ACTION OF EACH MUSCLE GROUP ON THE PREDICTED THRUSTLINE FOR THE SPINE AS AN ARCH

In order to determine the possible role of each individual muscle group towards the stability of the spine, the predicted thrustlines for the combination of each muscle group and body weight forces were compared with the predicted thrustline due to body weight alone.

4.3.1 MODEL PREDICTIONS FOR THE ERECT POSTURE

In comparison with the posterior convex curve due to body weight, the addition of large spine extensor muscle forces moved the thrustline anterior to the reference line (Figure 4-13). The effects of the internal oblique muscle were also considered in this category.

Figure 4-13: The predicted thrustlines moved anterior to the reference line due to extensor muscle action.
In contrast the addition of abdominal muscle forces rectus abdominis and external oblique considerably increased the overall thrustline curve in the posterior direction (Figure 4-14).

Figure 4-14: The predicted thrustlines increased in curvature in the posterior direction due to action of rectus abdominis and external oblique.
Some muscle groups resulted in a local flattening effect on the thrustline in the region of force application. Figure 4-15 shows this effect for muscle forces applied in the cervical region.

Figure 4-15: A local flattening effect on the thrustline in the cervical region due to muscle groups acting in these regions.

A similar effect was observed for the application of intra abdominal pressure in the lumbar region (Figure 4-16).

Figure 4-16: A local flattening effect on the thrustline in the lumbar region due to intra abdominal pressure forces.
Other muscle groups had very little effect on the thrustline curve. The predicted thrustlines are shown in Figure 4-17.

Figure 4-17: The predicted thrustlines for muscle groups with small effects
Finally some muscle groups increased the overall curve of the thrustline in the posterior direction, but to a smaller extent than the abdominal muscles (Figure 4-18).

Figure 4-18: The predicted thrustlines with increased curvature in the posterior direction due to muscle activity.

Accompanying the localised flattening effect on the thrustline in the cervical or lumbar regions, an overall reduction in the thrustline curve was observed, the extent of which varied for each muscle group. An overall classification of the muscle groups was made to compare the relative behaviour. A complete description and the accompanying predicted thrustlines are reported in the appendix (section 8.4):

**Category one:** Thrustlines moved significantly in the anterior direction.

**Category two:** Moderate movement of the overall thrustline.

**Categories three and four:** Local effects in the cervical region, with various degrees of flattening of the overall thrustline.

**Categories five and six:** Small and negligible movement of the overall thrustline respectively.

**Category seven:** Local effects in the lumbar region.

**Categories eight and nine:** Thrustline curve increased by various degrees.
4.3.2 ANALYSIS OF THE MUSCLE GROUP BEHAVIOUR ON THE PREDICTED THRUSTLINE FOR THE SPINE AS AN ARCH IN THE ERECT POSTURE

4.3.2.1 Muscle Groups Attaching to the Sacrum or Pelvis

Muscle groups with the most significant effect on the overall thrustline were the lumbar (including thoracic iliocostalis and longissimus thoracis which span this region) and abdominal muscle groups which attached to the spine or ribcage and the pelvis or sacrum. Due to the fixed representation of the pelvis, the forces generated at this end were not considered (see page 143). The muscle forces and their consequent moments generated at the spinal attachments are therefore unopposed.

The dominant moments generated by body weight forces alone were due to the parallel components in the thoracic and lumbar regions. In these regions, the applied force ranged from 10N to 31N. In contrast forces due to muscles in these regions range from 44N to 178N for longissimus thoracis, 108N to 189N for iliocostalis lumborum and 39N to 157N for lumbar multifidus. Similarly the abdominal muscle forces ranged from 200.5N (rectus abdominis) to 384.67N (internal obliques) generated by their individual fascicles. These muscle groups therefore generate significantly larger moments than body weight.
4.3.2.1.1 Lumbar Muscles

Figure 4-19 shows the moments generated by the components of combined body weight and muscle forces and forces due to intra abdominal pressure.

![Figure 4-19: Moments generated by the parallel and perpendicular components of force for body weight and lumbar muscle groups in the erect posture.](image-url)
For all lumbar muscle groups the dominant parallel component of force generates large moments about the ends of the spine (A and B). For longissimus thoracis, multifidus, iliocostalis and quadratus lumborum the parallel components of force act posterior to the reference line and generate moments which oppose flexion. In contrast due to lumbar lordosis, the points of application for the psoas muscle forces on the vertebral bodies and discs are positioned anterior and posterior to the reference line. The lines of action of these muscle groups are shown in Figure 4-20 and Figure 4-21 (longissimus and thoracic iliocostalis).

Figure 4-20: Illustrating the lines of action of forces generated by lumbar muscle fascicles.
For psoas the magnitude of forces acting anterior are (211N, 211N, 161N, 191N, 119N, 79N: total 893N) and of forces acting posterior are (61, 36, 101, 120, 173: total 491N). Due to similar moment arms for these anterior and posterior forces, the parallel components with the greater total magnitude anterior to the reference line generate the greater moments. These act with the moments generated by anterior body weight forces, increasing the overall flexion moment (Figure 4-19).
However, the greatest moments are generated about B by the perpendicular components of force for all lumbar muscles except longissimus thoracis. This is because these forces are mainly applied in the lumbar region with a large moment arm relative to the end B. This will affect the reaction force R1 which determines the initial projection of the thrustline. For all groups except thoracic iliocostalis and quadratus lumborum, the majority of forces have positive perpendicular components which generate moments opposing those due to body weight (Figure 4-19).

The total moment generated by the perpendicular and parallel components of force is shown in Figure 4-22. The combination of moments due to perpendicular and parallel components of varying magnitude results in some positive acting with body weight, and some negative acting against.

Figure 4-22: Total moments generated by the forces for body weight and lumbar muscle groups in the erect posture.
For longissimus thoracis moments due to perpendicular and parallel components of force oppose the direction of body weight and reverse the direction of the total moment generated about A and B. This consequently reverses the signs of the end support reactions.

Lumbar multifidus and iliocostalis reverse the total direction of moments about A, and thoracic iliocostalis and quadratus lumborum reverse the moments about B. These also affect the signs of the appropriate end support reactions (R2 and R1 respectively). For the remaining muscle groups a change in moments reflects an increase or decrease in the end support reactions, but they remain in the same direction as those due to body weight.

The values for the end support reactions for the lumbar muscle groups are reported in Table 4-5. For each muscle group, the generated moments result in different end support reactions and different initial thrustline projections.

Table 4-5: Calculated end support reaction components (N) for the applied lumbar muscle and body weight forces in the erect posture.

<table>
<thead>
<tr>
<th>Force Group</th>
<th>R1</th>
<th>R2</th>
<th>H1</th>
<th>H2</th>
<th>R1/H1</th>
<th>Initial Projection</th>
</tr>
</thead>
<tbody>
<tr>
<td>Body Weight</td>
<td>-14.32</td>
<td>22.54</td>
<td>379.87</td>
<td>-34.59</td>
<td>-0.0377</td>
<td>anterior</td>
</tr>
<tr>
<td>Psoas</td>
<td>-118.06</td>
<td>36.43</td>
<td>1831.51</td>
<td>-34.59</td>
<td>-0.06446</td>
<td>anterior</td>
</tr>
<tr>
<td>Lumbar multifidus</td>
<td>-84.08</td>
<td>-21.51</td>
<td>1083.95</td>
<td>-34.59</td>
<td>-0.07757</td>
<td>anterior</td>
</tr>
<tr>
<td>Iliocostalis lumborum</td>
<td>-29.93</td>
<td>5.83</td>
<td>1010.26</td>
<td>-34.59</td>
<td>-0.02963</td>
<td>anterior</td>
</tr>
<tr>
<td>Longissimus thoracis</td>
<td>11.33</td>
<td>-96.61</td>
<td>1948.15</td>
<td>-34.59</td>
<td>0.005816</td>
<td>posterior</td>
</tr>
<tr>
<td>Thoracic iliocostalis</td>
<td>59.57</td>
<td>15.93</td>
<td>922.66</td>
<td>-34.59</td>
<td>0.064563</td>
<td>posterior</td>
</tr>
<tr>
<td>Quadratus lumborum</td>
<td>57.15</td>
<td>15.25</td>
<td>959.6</td>
<td>-34.59</td>
<td>0.05956</td>
<td>posterior</td>
</tr>
<tr>
<td>Intra abdominal pressure</td>
<td>-68.13</td>
<td>12.02</td>
<td>379.46</td>
<td>-34.59</td>
<td>-0.1795</td>
<td>anterior</td>
</tr>
</tbody>
</table>
For longissimus thoracis, the initial projection of the thrustline determined by the positive value of R1 is posterior to the reference line. The addition of muscle forces with positive perpendicular and negative parallel components of force increases the perpendicular component and decreases the parallel component of the compression force vector, resulting in a positively increasing curve (Figure 4-23). Forces due to body weight with negative perpendicular and parallel components of force oppose this, although their effects are smaller due to lower force magnitudes. Shearing of this thrustline returns this curve anterior to the reference line (Figure 4-13).

![Diagram of thrustline and force vectors](image)

**Figure 4-23:** For longissimus thoracis, the initial projection of the thrustline is posterior to the reference line, and the local application of forces results in a positively increasing curve.

For lumbar iliocostalis and multifidus, the initial projection determined by the negative value of R1 is anterior to the reference line. The addition of forces with positive perpendicular components opposes the negative increasing curve due to body weight forces (Figure 4-24). The effects of the muscle forces dominate due to larger force magnitudes and final shearing of this thrustline maintains the projection anterior to the reference line (Figure 4-13).
A similar effect occurs for intra abdominal pressure whose forces oppose the negative perpendicular components of the body weight forces. The initial anterior projection of the thrustline is followed by a change in direction due to the intra abdominal pressure forces in the posterior direction in the lumbar region. However, the force magnitude is smaller than that applied by the lumbar muscles. A gradual negatively increasing curve therefore follows in the thoracic and cervical regions due to body weight forces (Figure 4-25). When sheared, the overall thrustline shows a flattening of the curve in the lumbar region (Figure 4-13).
Figure 4-25: For intra abdominal pressure a localised change in thrustline direction occurs.

For thoracic iliocostalis and quadratus lumborum, the initial projection of the thrustline is posterior to the reference line. Small negative components of force act with body weight, resulting in a negatively increasing curve which maintains its projection posterior to the reference line when sheared. However, due to the increased moments, the initial projection of the thrustline has a greater perpendicular component than that due to body weight alone. Gradual changes in thrustline direction at the addition of muscle and body weight forces are smaller resulting in a smaller depth of curvature. A small overall flattening effect on the thrustline due to body weight therefore occurs.
Figure 4-26: For thoracic iliocostalis and quadratus lumborum the initial projection of the thrustline is posterior to the reference line, but the addition of muscle forces results in a negatively decreasing curve.

For psoas, the initial projection is anterior to the reference line but has a greater perpendicular component (Figure 4-24). The muscle forces have positive and negative perpendicular components of force, which due to opposing terms reduces the effects of force application. However, due to the greater perpendicular component of the initial thrustline portion, a steeper curve than that due to body weight alone is obtained.
4.3.2.1.2 Abdominal Muscle Groups

For the internal oblique muscle due to positive perpendicular components of force, moments are generated which act against body weight (Figure 4-27). In contrast rectus abdominis and external oblique act with body weight. For all three muscle groups the parallel component of force acting anterior to the spine increases the flexion moment. This is increased considerably due to the magnitude of the forces, and the large moment arm.

![Diagram showing moments generated by the parallel and perpendicular components of force for body weight and abdominal muscle groups in the erect posture.](image)

Figure 4-27: Moments generated by the parallel and perpendicular components of force for body weight and abdominal muscle groups in the erect posture.
The addition of moments due to parallel and perpendicular components of force of different magnitudes and sign results in different direction for the overall moment generated about A and B (Figure 4-28). The moments due to rectus abdominis and internal oblique act with body weight and increase the values about B, thereby increasing R1. In contrast the moments about B due to the external oblique muscle are of greater magnitude but oppose the direction of those due to body weight, thereby reversing the sign of R1. The thrustlines are therefore initially projected in different directions. The predicted values for the end support reactions are shown in Table 4-6.

Figure 4-28: Total moments generated by the forces for body weight and abdominal muscle groups in the erect posture.
Table 4-6: Calculated end support reaction components (N) for the applied abdominal muscle and body weight forces in the erect posture.

<table>
<thead>
<tr>
<th>Force Group</th>
<th>R1</th>
<th>R2</th>
<th>H1</th>
<th>H2</th>
<th>Initial Projection</th>
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<tr>
<td>Body Weight</td>
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<td>379.87</td>
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<td>1168.73</td>
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<td>1538.15</td>
<td>-34.59</td>
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<td>448.77</td>
<td>1573.00</td>
<td>-34.59</td>
<td>posterior</td>
</tr>
</tbody>
</table>

For rectus abdominis which acts in the direction of body weight forces the initial projection of the thrustline is anterior to the reference line. Due to the magnitude of these forces a steeper thrustline curve results.

For the external oblique muscles, due to the positive value of R1 the initial projection of the thrustline is posterior to the reference line. The addition of forces due to body weight and muscles with negative perpendicular and parallel components result in thrustline with positive decreasing and eventually increasing negative gradient. The thrustline is therefore sheared posterior to the reference line. Due to the magnitude of the muscle force, a greater change in direction at each portion of the thrustline, and an overall steeper curve is generated.

In contrast for internal oblique, the applied forces with positive perpendicular components act against body weight reducing the slope of the thrustline curve. As for lumbar multifidus and iliocostalis (Figure 4-24) the thrustline follows an anterior convex curve.
4.3.2.2 Effects on R2

For the lumbar and abdominal muscle forces, the value of R2 is also affected. For body weight, moments due to parallel components of force increase the value of R2 (page 263). For psoas, rectus abdominis and external oblique which have parallel components of force anterior to the spine, the values of R2 are increased further. In contrast for the remaining lumbar muscle groups, the parallel components of force act posterior to the reference line, and generate moments in the opposite direction as those due to body weight. For thoracic iliocostalis, quadratus lumborum and intra abdominal pressure, the value due to body weight is reduced. For longissimus thoracis, lumbar iliocostalis and lumbar multifidus, the value is of opposite sign and increased. This indicates more posterior force than required.

4.3.2.3 Muscle Categorisation

The greatest effects on the thrustline were observed for longissimus thoracis, lumbar iliocostalis and multifidus which resulted in a thrustline anterior to the reference line. These were therefore included in the category one. Also large effects were observed for psoas, although this resulted in a thrustline posterior to the reference line and was therefore classified in category eight. The direction of force for thoracic iliocostalis and quadratus lumborum, distinguished them from the other lumbar muscle groups by having only a small effect on the predicted thrustline. These were included in categories six and five respectively. Finally, the abdominal muscles with the greatest overall increase in thrustline curve were included in category nine.
4.3.2.4 Muscle Groups Acting Above the Sacrum and Pelvis

Muscle groups acting above the sacrum and pelvis generate pairs of forces in opposing directions at their attachment points. Those attaching between the vertebrae and head generate forces above and below the end B. Those attaching between the vertebrae or between the vertebrae and ribcage generate forces between the ends A and B. This is illustrated in Figure 4-29.

For a muscle with an oblique orientation, the attachment points differ in their positioning relative to the reference line. The force generated at the upper attachment in Figure 4-29 (either between A and B or above and below B) is positioned further posterior to the reference line, so that the parallel component has a moment arm greater than that at the lower attachment. Despite forces of equal magnitude, the moment due to parallel components of force is therefore greater at the upper attachment.
For the perpendicular components, when forces act between A and B equal moments are generated in opposing directions. However, when the upper attachment is above B, the moments generated are in the same direction about B. Therefore despite forces of opposing directions, small differences in moments occur.

The direction of these moments depends upon the orientation of the muscle fascicles for each group. Figure 4-30 shows the lines of action of forces generated by the larger cervical muscle groups attaching to the head. The attachment of semispinalis capitis and spinalis capitis to the head is further posterior to the reference line than their vertebral attachments. In contrast the attachments to the vertebrae for splenius capitis are positioned further posterior than their attachments at the head.

Figure 4-30: Illustrating the lines of action of forces due to large cervical muscles attaching to the head.
Figure 4-31 shows the moments generated by parallel and perpendicular components of force for these cervical muscle groups. The parallel components of force generated by semispinalis capitis and spinalis capitis at their upper attachments have larger moment arms and act in a negative direction posterior to the reference line. Negative moments are therefore generated about the ends A and B which oppose those due to body weight. In contrast the parallel components of splenius capitis generated at the lower attachments in an upwards direction have a larger moment arm and increase the positive moments generated by body weight alone.

![Figure 4-31: Moments generated by the parallel and perpendicular components of force for body weight and large muscle groups attaching between head and spine in the erect posture.](image-url)
Also, the perpendicular components of force due to semispinalis and spinalis capitis are negative above B and positive below B, thereby generating a couple about B in a positive direction. In contrast perpendicular components of force due to splenius capitis generate a couple about B in a negative direction.

Smaller differences in moments are observed for longissimus capitis due to force components with positive and negative perpendicular direction, and for longus capitis due to a less oblique orientation. Also the magnitude of the forces and the moments they generate are smaller.

Figure 4-32: Total moments generated by the forces for body weight and large cervical muscle groups attaching to the head in the erect posture.
Despite this variation in moments, the addition of moments due to perpendicular and parallel components results in a very small difference in the total moment in comparison with those due to body weight and these differences are hardly visible (Figure 4-32). Similar results were found for the remaining muscle groups acting above the sacrum and pelvis. Consequently, only small changes in the end support reactions (R1 and R2) occur. The effect on the thrustline can therefore be explained mainly by the local application of forces.

4.3.2.4.1 Force Application

For muscles with attachments above the sacrum, the forces generated in opposing directions ensure equilibrium in the direction parallel to the reference line without an increase in H1. Also the generated moments for these muscle forces have very little effect on the end support reaction components R1 and R2. The projection of the thrustline due to body weight is therefore followed up until the application of these muscle forces in a particular region.

The muscle forces generated at the lower attachments act upwards with positive parallel components which oppose the effect of body weight and reduces the gradient of the following thrustline sections. Positive perpendicular components of force will also oppose the effects of body weight, whereas negative perpendicular components will act with body weight reducing the effect of the parallel component. The application of a muscle force at the upper attachment opposes the direction of the initial muscle force, but does not counteract the flattening effect on the thrustline over several levels. Consequently, the overall depth of the thrustline curve is reduced. Muscles of this type have a greater effect with the number of levels spanned.
4.3.2.4.2 Muscle Categorisation

4.3.2.4.2.1 Cervical Muscles Between the Vertebrae and Head

For muscles attaching between the head and the vertebrae, the forces acting above B are incorporated into H2 which affects the direction of the final portion of the thrustline. However, because this portion is very small between C1 and C2, the forces applied at the lower attachments of these muscles have a more significant effect.

The forces generated at the lower attachment for semispinalis capitis, longissimus capitis and spinalis capitis have positive perpendicular and parallel components which act against body weight. A change in thrustline direction therefore occurs in the cervical region due to the application of these muscle forces. This is illustrated in Figure 4-33. Application of forces due to semispinalis capitis of large magnitude over a number of levels in the thoracic and cervical regions results in a considerable flattening effect on the thrustline in these regions as observed (Figure 4-13). The effects of this muscle were therefore classified as category one, moving the overall thrustline anterior to the reference line.
Figure 4-33: Illustrating the effects of cervical muscle forces on the thrustline initially determined by the action of body weight (not to scale).

Categories two, three and four describe the localised effect in the cervical region to various degrees.

Spinalis capitis generates two forces of relatively large magnitude (100N each) at the C6 vertebral level. These forces have both positive perpendicular and parallel components, and therefore result in a considerable flattening of the thrustline in the cervical region, and of the overall thrustline. This muscle was therefore included in category three. The abrupt change in thrustline direction observed in Figure 4-15 is due to the application of this force at a single level (C6).
Muscle groups longissimus capitis and longus capitis generate forces of similar magnitude to each other (80N and 92N respectively), but in different directions. At the vertebral attachments longissimus capitis generates forces with small positive and negative perpendicular components where longus capitis generates forces directed anteriorly and the perpendicular components are negative. The flattening effect due to longissimus capitis is therefore greater and this muscle was included in category three, although the flattening effect is not as large as that due to spinalis capitis due to forces of smaller magnitude. The lesser effects of longus capitis were included in category four.

Forces due to splenius capitis were of much greater magnitude (four forces of 106.5N each) but at their vertebral attachments are oriented with negative perpendicular components of force acting with body weight. This reduces the localised flattening effect of the upward parallel component of forces, and only a small overall effect is visible. Hence despite its large magnitude this muscle groups was included in category five.

Forces due to muscles attaching between the head and the upper two vertebral levels also show small localised flattening effects in the cervical region. However, their forces are effective only for these end vertebral levels. Small changes in thrustline curvature in the cervical region and overall are thus observed for these muscles.

Rectus capitis posterior major and minor generated forces at their lower attachments with positive perpendicular and parallel force components and their resulting flattening effect was included in category four (Figure 4-34).
The single force generated by obliquus capitis superior is of greater magnitude and has greater perpendicular components than those due to rectus capitis posterior but is applied at the transverse process of the C1 vertebra which is very close to the end of the spine. The force therefore has little effect. Consequently, obliquus capitis superior was classified in category six.

Rectus capitis lateralis and anterior generate relatively small forces (21N and 13N respectively). They were therefore considered to have very little effect on the overall thrustline and were also classified as category six.

![Diagram](image)

Figure 4-34: Illustrating the lines of action of forces due to small cervical muscles attaching between the spine and the skull.
4.3.2.4.2.2 Cervical Muscles Between Vertebrae

Muscle groups longus coli, longissimus cervicis, splenius cervicis and semispinalis cervicis which attach at both ends to the cervical vertebrae are all of similar relative magnitude (cross sectional area 81-120 mm$^2$). These are of smaller magnitude than those muscles attaching to the head in category three and the local flattening effect is smaller. Consequently, these are all included in category five. The lines of action of these muscles are illustrated in Figure 4-35.

Fascicles of splenius cervicis and longissimus cervicis act over a greater number of vertebral levels (Figure 4-35). However, these also show the greatest oblique orientation and generate force with negative perpendicular components at their lower attachments. Therefore only the parallel components of force act against body weight. In contrast forces due to semispinalis cervicis follow a line of action from the spinous processes at upper levels to the transverse processes at lower levels. Although the spinous processes have greater posterior positioning relative to the vertebrae, the thoracic curvature results in the transverse processes at approximately equal positioning relative to the reference line. The perpendicular components of these forces are therefore less, which compensates for the fewer number of spinal levels which they span. Consequently, these muscle groups all have a similar effect on the thrustline and are included in category five.

In contrast spinalis cervicis has a larger cross sectional area (240 mm$^2$) and generates larger positive perpendicular force components at its lower attachments. These forces therefore have a greater flattening effect on the thrustline and the muscle was included in category three.
The smaller muscle obliquus capitis inferior (Figure 4-34) has attachment points at the C1 spinous process and C2 transverse process positioned very close together. Consequently, the effect at adjacent levels is very small. This muscle was therefore included in category six.

4.3.2.4.2.3 Thoracic Region Muscles Between Vertebrae

For spinalis thoracis, both upper and lower attachments are at the spinous processes, and are of approximate equal positioning on either side of the convex thoracic curve resulting in forces almost parallel to the reference line (Figure 4-21). The total force due to spinalis thoracis and semispinalis thoracis is 105 N and 58N respectively. However, these forces act respectively over nine and twelve of the thoracic levels of the spine, so that the individual forces applied at each level are no more than 12.5N, and the consequent effects on the overall thrustline are very small. This small flattening effect occurs mostly in the thoracic region where forces are applied. These were therefore included in category six.
4.3.2.4.2.4 Intersegmental Muscles

Forces due to intertransverse and interspinalis muscle groups follow the spinal curve in the cervical, upper and lower thoracic, and lumbar regions. This is illustrated in Figure 4-36.

Figure 4-36: Illustrating the lines of action of forces due to intersegmental muscles.
The muscles attached between vertebrae at the beginning of lordotic cervical and lumbar curves are oriented anteriorly and generate forces at the lower attachments with negative perpendicular components. At the ends of the lordotic curves the vertebrae are oriented posteriorly, and for intertransverse the forces generated at the lower attachments have positive perpendicular components. Due to smaller dimensions for the spinous processes at the upper cervical levels, interspinalis muscle forces still have negative perpendicular components. Consequently, intertransverse has a slightly greater flattening effect than interspinalis despite a smaller magnitude. This can be seen in figure 8-8 in the appendix.

For either case, the dominant parallel component of force at the lower attachment is positive. A small flattening effect occurs between the application of the opposing force pairs, resulting in a gradual flattening in the cervical and lumbar regions. Forces with positive perpendicular components enhance this effect, whereas forces with negative perpendicular components reduce the effect. The magnitude of the applied forces is greater than the body weight forces in the cervical regions, resulting in a more visible flattening effect for this region. These muscle groups were therefore included in category four.

For rotatores the attachments to vertebral laminae results in more oblique forces with greater perpendicular components of force (Figure 4-36). The forces at the lower attachment points have negative perpendicular components which act with body weight, but positive parallel components which oppose body weight forces. The greater perpendicular components for rotatores results in a smaller reduction in slope of the thrustline than that due to interspinalis or intertransverse. Also this occurs only in the thoracic region. The effects of rotatores on the overall thrustline are therefore less (Figure 4-17), and this muscle was included in category six.
Forces due to cervical and thoracic multifidus also act in pairs between adjacent vertebrae, although due to their attachment between spinous and articular processes the perpendicular component of force is dominant. For the initial two force pairs in the thoracic region, and a single force pair in the cervical region, the force applied at the lower attachment has positive perpendicular and parallel components (Figure 4-36). However, for the remaining forces curvature of the spine alters the position of the attachment points and the perpendicular components are negative at the lower attachments. Due to the dominant perpendicular components these forces act with body weight, resulting in a small increase in the thrustline gradient at each section. The overall effect is thus an increase in the steepness of the thrustline curve, and this muscle was classified in category eight.

4.3.2.4.2.5 Muscle Groups Attaching Between the Vertebrae and the Ribcage

Scalenus posterior and iliocostalis cervicis both attach between the vertebrae and the ribcage and have an oblique orientation. The force generated at the lower attachment has negative perpendicular and positive parallel components. These perpendicular components are greater than those due to body weight in this posture, and result in an increase in thrustline curvature. The opposing force generated at the upper end with positive perpendicular component opposes this effect indicated by a small reduction in thrustline slope shown in Figure 4-18, although between these points of attachment a small increase in thrustline occurs.
The transverse abdominis muscle which attaches between the spinous processes and the linea alba (positioned anterior to the spine) also generates forces with dominant negative perpendicular components and small positive parallel components at the lower attachments. However, due to the magnitude of these forces applied, the effect on the overall thrustline more noticeable (Figure 4-16).

The application of forces generated at the lower attachments increases the negative slope of the thrustline due to body weight. However, this is followed by a decrease in negative thrustline slope with the addition of forces at the upper attachments with positive force pairs. Finally the forces due to body weight continue in the thoracic and cervical regions to follow a negatively increasing curve. This is illustrated in Figure 4-38. The overall thrustline thus shows a localised flattening in the lumbar regions. For the ribcage muscles the change in direction due to the application of positive perpendicular forces at the upper attachments was less visible due to smaller force magnitude.
Figure 4-38: Illustrating the effects of transverse abdominis muscle forces on the thrustline initially determined by the action of body weight (not to scale)
4.3.3 MODEL PREDICTIONS FOR MUSCLE GROUP BEHAVIOUR WITH FLEXION

Due to the increase in steepness of the thrustline curve predicted for body weight forces with flexion, the effects of each muscle group were generally reduced in flexion. However, for some muscles the relative strengths were increased. Significant changes were reflected in movement of the muscle groups between categories. Table 4-7 summarises the muscle groups belonging to each of the nine categories for the four postures investigated and shows this relative movement. The predicted thrustlines for each category for each of the three flexed postures can be found in the appendix (section 8.4).

The table is arranged in four sections to show relative movement of the muscle groups between categories. This includes cervical muscles (first 11 rows), lumbar muscles (rows 13-17), muscles with little effect in all four postures (rows 19-29) and muscles which increase the thrustline in the erect posture (rows 31-37).
Table 4-7: Showing the categories for each muscle group for the four postures investigated.

<table>
<thead>
<tr>
<th>Row Number</th>
<th>Muscle Group</th>
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<th>Flex2</th>
<th>Flex3</th>
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4.3.3.1 Relative Movement Between Categories

In the flex1 posture, the absence of lumbar muscles iliocostalis lumborum and multifidus from category one shows a reduction in their effectiveness in flattening the overall thrustline. These move to category seven in the flex1 posture in which only a localised flattening of the lumbar region is significant. This also occurs for longissimus thoracis in the flex3 posture. Also internal oblique is included in category one for the erect posture, but increases the curvature of the thrustline in flexion, moving to category nine in the flex2 and flex3 postures.

For some muscles an increase in effectiveness occurs with flexion. Longissimus capitis moves from category three in the erect posture, to category two in the flex1 and flex2 postures and to category one in the flex3 posture. The localised flattening effect of this muscle in the cervical region is less noticeable than its increase in the effectiveness to flatten the overall thrustline curve in flexion. Similar movement is observed for splenius cervicis and splenius capitis initially in category five for the erect posture. Also longissimus cervicis, semispinalis cervicis and longus colli in category five for the erect and flex1 posture move to category two in the flex3 posture.

The muscles scalenus posterior and iliocostalis cervicis which attach between the spine and the ribcage result in an increase in thrustline curve in the erect posture. With flexion, this role reverses and a flattening of the thrustline occurs. These muscle groups thus move from category eight in the erect posture, to category six in the flex1 and flex2 postures and to category five in the flex3 posture.
For some muscles a decrease in effectiveness occurs initially with flexion, followed by an increase. For example intertransverse, interspinalis and longus capitis moves from category four in the erect posture to category five for the flex1 and flex2 postures and finally to category three in the flex3 posture (category 2 for interspinalis).

For the remaining muscle groups the relative effects on the thrustline are maintained. Muscles such as rotatores, rectus capitis lateralis and anterior, spinalis and semispinalis thoracis and obliquus capitis superior remain in category six and have a very small effect on the thrustline. Also spinalis capitis remains in category three and semispinalis capitis remains in category one and psoas remains in category eight for all four postures. Similarly the relative effects of the abdominal muscle groups rectus abdominis and external oblique in category nine were maintained.

4.3.4 ANALYSIS OF THE EFFECTS OF FLEXION ON THE ACTION OF EACH MUSCLE GROUP ON THE PREDICTED THRUSTLINE

4.3.4.1 Lumbar Muscles

During flexion the magnitude of the force each muscle group exerted is not changed. However, the relative proportion of force in the perpendicular and parallel directions relative to the reference line is affected. This is due to the increased inclination of the reference line and a change in orientation of the muscle line of action accompanying a change in relative movement of the vertebrae to which they are attached.
For the lumbar muscles, the increase in convex curvature of the spine results in a change in force direction relative to the reference line. Figure 4-39 compares the direction of the lumbar muscle forces (longissimus thoracis and lumbar iliocostalis and multifidus) relative to the reference line in the erect and flex3 postures. The downward forces close to 180 degrees in the erect posture increase their perpendicular direction in the negative direction in the flex3 posture and are oriented less than 180 degrees.

Figure 4-39: Comparison of the directions of lumbar muscle forces in the erect and flex3 posture.
Figure 4-40 shows the general change in direction of forces in the lumbar regions which accompanies flexion. Muscles attached between the lumbar vertebrae and pelvis follow a direction almost parallel to the reference line in the erect posture. With flexion, the attachment points on the vertebrae are positioned further posterior from the reference line and from the pelvis, so that their direction relative to the reference line becomes more oblique. The negative component of force is consequently increased.

![Diagram showing the change in direction of lumbar and cervical muscle forces with flexion.]

**Figure 4-40:** Illustration of an increase in the direction of lumbar and cervical muscle forces in the negative perpendicular direction relative to the reference line due to flexion.

However, although the components of the muscle forces are increased in the negative direction with flexion, the perpendicular components of body weight are also increased. The lumbar muscle forces therefore still oppose the effects of forces due to body weight.
However, the effects of the moments generated by the lumbar muscle forces are reduced with flexion. In the erect posture the moments generated by the lumbar muscles in category one dictated the projection of the thrustline anterior or posterior to the reference line. However, with flexion the perpendicular components of force due to body weight increase and the moments generated by muscle forces have smaller effects.

In the flex2 posture, the increase in negative perpendicular components of the muscle forces results in moments which act with those due to body weight about A and B. This is illustrated in Figure 4-41 where only the moments due to intra abdominal pressure result in total moment less than that due to body weight about A and B. Also the moments due to the parallel components of muscle forces result in an increase in total moment in the same direction as those due to body weight only. Thus in contrast to the erect posture, the total moment due to lumbar muscle perpendicular and parallel components is therefore in the same direction as that due to body weight. The projection of the thrustline therefore follows the direction of that due to body weight, posterior to the reference line. Similar findings occur for the flex3 posture.
4.3.4.2 Abdominal Muscles

The internal oblique muscle included in category one in the erect posture significantly increased the thrustline curve in the flexed postures. In the erect posture, the perpendicular components of force act against body weight and reverse the overall projection of the thrustline. However in the flexed postures, the perpendicular components of body weight increase in the negative direction. The muscle forces are therefore less effective in opposing the effects of body weight and the overall projection of the thrustline is posterior to the reference line as for body weight. Also the generated moments due to the parallel component of this muscle force contribute to the increase in end support reactions and an increase in thrustline curve.
4.3.4.3 Cervical Muscles

For the cervical muscles an increase in convex curvature of the spine with flexion also results in a change in force direction relative to the reference line. Figure 4-42 compares the direction of the cervical muscle forces relative to the reference line for the erect and flex3 postures. With flexion, these forces move closer to 360 degrees or beyond, indicating a reduction in positive perpendicular component or an increase in the negative direction.

These forces attaching between the head and the vertebrae are affected by the increase in curvature. Again this is due to an increase in posterior positioning of the vertebral attachments with flexion, illustrated in Figure 4-40. This directs the forces towards the head with greater negative perpendicular force components.

Figure 4-42: Comparison of the directions of cervical muscle forces in category one in flex3 posture with the directions in the erect posture.
4.3.4.4 Intersegmental Muscles

For interspinalis and intertransverse which follow the curvature of the spine, the direction of forces are changed significantly with a reduction in lumbar lordosis. The vertebrae of the lumbar and lower thoracic regions are directed posteriorly, so that the forces generated at the lower attachment point of each muscle pair have positive perpendicular components. In contrast, the vertebrae of the cervical region are oriented anteriorly and result in negative perpendicular forces at the lower attachment point of each muscle pair (Figure 4-43). This results in an increase in effectiveness on flattening the thrustline in the lumbar region and a decrease in the cervical region.

![Figure 4-43: Forces generated at the lower attachment point of each interspinous and intertransverse force pair have positive perpendicular components in the lumbar region and negative perpendicular components in the cervical region for a flexed configuration.](image-url)
For multifidus, significant changes in force direction occur in the cervical region due to flexion. In the erect posture at upper cervical levels, the force generated at each lower attachment of the force pair is directed upwards and with negative perpendicular components. For the flex2 posture, the positioning of the points of attachment relative to the reference line are altered and the same forces are directed upwards with positive perpendicular components. This is illustrated in Figure 4-44. Thus in the erect posture, the negative components of force act with body weight whereas in the flexed postures the positive components act against body weight. Due to the dominance of these perpendicular components the effect is significant and the thrustline curve is increased in the erect posture but flattened in the flexed postures.

![Figure 4-44: Comparing the lines of action of forces due to cervical and thoracic multifidus in the erect and flex2 postures.](image)
4.3.4.5 Rib Cage Muscles

For scalenus posterior and iliocostalis cervicis the oblique line of action between the ribcage and the vertebrae is maintained with flexion. However, the perpendicular components of body weight forces are increased and the parallel components decreased. Relative to the body weight forces, the proportion of the parallel components for the muscle forces is increased with flexion. These muscles therefore have a flattening effect on the thrustline which increases with flexion as the perpendicular component of body weight force increases. This is illustrated by their movement from category eight in the erect posture to category six in the flex1 and flex2 postures to category five in the flex3 posture.

4.3.4.6 Relative Movement between Cervical and Lumbar Muscle Groups

In the erect posture the greatest effects on the thrustline due to body weight were observed for lumbar muscles which generate large forces and moments. However, in flexion the effects of these moments are diminished and the relative effects of the lumbar muscles on the overall thrustline are reduced. In contrast, the effects of some cervical muscles (longissimus capitis, splenius capitis and splenius cervicis) are increased in the flex3 posture.

Both the lumbar and cervical muscles are observed to increase their perpendicular components of force in the negative direction with flexion. The effectiveness of these muscles in opposing the directions of the body weight forces is thus reduced. However, the lumbar muscles still generate greater force than the cervical muscles.
The increase in effectiveness of some cervical muscles groups relative to that of the lumbar muscles may be due to the difference in downward and upward directions of force for these muscles. In both cases the muscle forces are not as oblique as the body weight forces in flexion. The slope of the resultant compression force vector is therefore reduced by the addition of muscle forces. For the lumbar muscles, negative perpendicular and parallel components of force are added to the compression force vector at each stage of force application. In contrast, for the cervical muscles negative perpendicular but positive parallel components are added, increasing the parallel component of the compression force vector. The effect of the cervical muscles is therefore significant, despite their smaller magnitude.
4.3.5 DISCUSSION

The contribution of each muscle group to the stability of the spine can be appreciated using the categories described. Lumbar and abdominal muscles in categories one and nine had a considerable effect on the thrustline, whereas muscles in categories five, six and eight had smaller effects. Also, muscles in categories two, three, four and seven were significant for their local effects on the thrustline path.

The bending moment on an arch is indicated by the distance between the predicted thrustline and centre line of the spine. The distance between the thrustline due to body weight projected anteriorly and the centre line of the vertebral bodies indicates a positive (flexion) moment on the arch.

The large extensor muscles of the spine and of the head (semispinalis capitis) resulted in a posterior projection of the thrustline and a negative distance between the thrustline and the spine in the erect posture. The overall moment on the spine was therefore an extension moment as expected. These muscles were considered essential in the flexed postures to support the weight of the upper body and additional external loads.

The thrustline curve due to body weight was increased by the abdominal muscles rectus abdominis and external oblique in category nine, indicating an increase in the flexion moment on the spinal arch. These muscles are therefore considered to oppose the effects of the extensor muscles in moving the thrustline closer to the spinal configuration.
Muscles of category five and six which had smaller effects on the thrustline due to body weight have little effect on the bending moments or on stability of the spine as a whole. These muscles may be more important for stability of the individual intervertebral joints. This included the small muscles attaching between the head and the atlas and axis whose contribution towards the stability of the atlanto-occipital joint may be more significant.

Muscles applied in the cervical region and forces due to intra abdominal pressure in the lumbar region were found to result in small localised flattening effects on the thrustline in their region of application. These are considered important for ensuring stability of the spine in postures which include cervical and lumbar lordosis.

Also along with a localised flattening effect, these muscles resulted in an overall decrease in thrustline curve depth. Contraction of these muscles to ensure regional stability may therefore also affect the overall stability of the whole spine. In particular, muscles contracting to ensure equilibrium of the head such as semispinalis capitis, or longissimus capitis, can significantly affect the thrustline path and dictate the forces required for stability at other levels.

The small increase in thrustline curvature due to scalenus posterior and iliocostalis cervicis indicated an increase in flexion moment at these levels, which was expected due to their roles in flexion. In the upright posture the action of the cervical and lumbar muscles to ensure stability in these regions is accompanied by an overall decrease in thrustline curve. The thoracic muscle forces may therefore be necessary for controlling the thrustline curve in the thoracic region and ensuring it corresponds to the convex curve of the vertebrae at these levels.
Contrary to expectations, a flattening effect on the thrustline occurred for longus coli indicating a small decrease in flexion moment whereas this muscle is known to flex the trunk. However, flexion occurs when one end of the muscle is fixed (the joint is controlled by several other muscles) whereas this model considered muscle forces pulling both vertebral attachment points together.

For the lumbar muscles a diminishing role with flexion was observed to an extent where the cervical muscles of smaller magnitude had greater effects on the thrustline. This was considered due to their straight line representation. The study showed force direction to be a more sensitive parameter than force magnitude. In particular in the erect posture forces of 25N due to intertransverse had a greater effect than forces of 50N due to interspinalis and four forces each of 106.5N for splenius capitis had less effect than two forces of 100N for spinalis capitis due to different directions. The straight line representation in which the direction tended towards an increase in negative perpendicular components may therefore not be suitable for investigating muscles which span the spinal curve. This was considered in the sensitivity analysis in section 4.5.
4.4 THE EFFECTS OF LIGAMENTS ON STABILITY OF THE SPINE

Testing of the effects of forces generated by the ligaments on the predicted thrustline was considered necessary so that the contribution of active muscles and passive ligament elements towards stability of the spine could be compared.

4.4.1 MODEL PREDICTIONS

Figure 4-45 shows the predicted thrustlines for ligament forces generated at points A and B of the characteristic load deformation curve previously described. For the posterior longitudinal ligament, very little effect on the predicted thrustline due to body weight is observed, and would be comparable with muscle groups in category six. Forces due to ligamentum flavum result in a slightly greater flattening effect, with small localised effects in the cervical and lumbar regions. This could be classified in category four along with the effects of the intersegmental muscles. A very small flattening effect in the lumbar region was observed when the number of vertebral levels spanned by the lumbar posterior longitudinal ligament was increased to two (Panjabi, 1991c, see page 239). However, the forces from both ligaments are insufficient to ensure stability at any level.
Figure 4-45: The effects of individual ligament group forces on the predicted thrustlines due to body weight for the erect posture.

Figure 4-46 shows the predicted thrustlines for the ligament groups and body weight in the flex3 posture. Again the effects of the posterior longitudinal ligament are small. Forces corresponding to points B and C on the characteristic force-deformation curve show little difference between them, although a greater effect results than for forces at point A. Similar observations are made for ligamentum flavum, although the effects of all force sets are greater than those for the posterior longitudinal ligament. However, although the ligamentum flavum has a greater flattening effect on the thrustline, this is comparable with muscle groups of category five. The effects of increasing the number of vertebral levels spanned by the lumbar posterior longitudinal ligament to two (Panjabi, 1991c, see page 239) had a very small flattening effect (Figure 4-46). Similar findings were found for the flex1 and flex2 postures.
4.4.1.1 Discussion

The posterior longitudinal ligament and the ligamentum flavum were reported by Hukins et al. (1990) to have the greatest force generating potential of all the ligament groups in the sagittal plane. However, in comparison with other muscle groups, these ligaments were respectively classified in categories four and six in the erect posture, and categories five and six in the flexed postures. Due to the very small effects of the posterior longitudinal ligament, the combination of this ligament with ligamentum flavum would show little difference to that observed for the ligamentum flavum alone. The combination of these two main ligaments therefore cannot support the spine in the postures tested in the sagittal plane.

Figure 4-46: The effects of individual ligament group forces on the predicted thrustlines due to body weight for the flex3 posture.
The effects of interspinalis and intertransverse muscles in the erect, flex1 and flex2 postures corresponded with the effects of ligamentum flavum. However, in the flex3 posture, the intersegmental muscles had a greater effect, with interspinalis moving to category two, and intertransverse moving to category three. In comparison the posterior longitudinal ligament had very little effect for all postures, and was included in category six.

Similar to the intersegmental muscles, the ligament fascicles are attached between the vertebrae so that the direction of their forces corresponds to the curvature of the spine. The ligaments also act between the vertebral bodies closer to the centre line of the spine at every level, and the applied forces were significantly greater than the forces assigned for the intersegmental muscles. A thrustline closer to the path of the vertebral bodies would therefore be expected.

However, the points of application of pairs of opposing forces are positioned closer between the vertebral bodies than the attachments on the transverse and spinous processes for the intersegmental muscles. The pairs of ligament forces therefore act over smaller intervals of the spine, and the altered direction of the thrustline path is effective over a shorter distance.

In particular, the posterior longitudinal ligaments are modelled as force pairs acting on the vertebral bodies level with the superior and inferior surfaces of the disc. The short distance between these points accounts for the small effects which are observed on the predicted thrustline despite the relatively large forces applied. Increasing the number of levels spanned to two in the lumbar regions had only a small flattening effect. For the ligamentum flavum a greater distance between the attachment points representing the vertebral laminae along the entire length of the spine allowed a greater flattening effect.
The inability of the ligament forces to support the spine in the postures investigated in the sagittal plane indicates the dominance of the spinal muscle groups in stability. However, the passive forces of the ligaments may provide a smaller contribution to the stability of the spine at the individual joints with minimal energy expenditure. This may be important for efficient functioning of the spinal system.
4.5 SENSITIVITY ANALYSIS

Due to sparsity of data regarding the forces of the individual fascicles of each muscle group, certain assumptions were made.

1) The direction of muscle forces was assumed to act in a straight line between the points of attachment.

2) Uniform force distribution was assumed for muscles for which only the total cross sectional area was reported.

3) The magnitude of force generated by the intersegmental muscles was assumed to be small. Due to lack of data for cervical and thoracic multifidus, and rotatores, arbitrary values were assigned.

However, the magnitude, points of force application and direction of a muscle have been shown to have significant effects on the predicted thrustline. It was therefore considered necessary to investigate the effects of these assumptions by considering a range of alternatives.
4.5.1 EFFECTS OF CURVATURE ON MUSCLE FORCE DIRECTION

For the majority of muscle groups, the line of action was assumed to be a straight line between the points of attachment. However, depending upon configuration, the curvature of the spine may cause the line of action of some muscles to deviate from this (see page 146). Muscle groups considered to be most affected by the flexed curvature of the spine were those muscles which are attached to the vertebrae at both ends and those which are attached to the vertebrae at one end and also span several segments (e.g. longissimus thoracis). The effects of curvature on these muscles was investigated by comparing their effects on the predicted thrustline due to body weight for straight and curved representations.

4.5.1.1 Comparison of the Predicted Thrustlines for Muscles with Straight or 'Curved' Force Representation

For each case considered, the greatest flattening effect was observed at the peak of the thrustline curve. The peak x co-ordinate of each thrustline due to muscle activity was therefore determined (relative to the reference line) and expressed as a percentage of the peak co-ordinate of the thrustline predicted for body weight alone. This was termed the ‘peak percentage’ and was used as a comparative measure for the effects of each force distribution. The change in peak percentage due to the curved representation was determined for each muscle group in each posture, and the values shown in Table 4-8. A negative sign indicates the distance between the thrustline and the reference line is decreased, indicating a reduction in curve depth.
Table 4-8: Change in peak percentage values due to curved representation of muscle forces. ⇔ indicates the peak thrustline position changes from posterior to anterior to the reference line.

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>Change in peak percentage from straight representation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Erect</td>
</tr>
<tr>
<td>Longissimus thoracis</td>
<td>0</td>
</tr>
<tr>
<td>thoracic iliocostalis</td>
<td>-119.1 ⇔</td>
</tr>
<tr>
<td>longissimus cervicis</td>
<td>-17.5</td>
</tr>
<tr>
<td>splenius cervicis</td>
<td>-1.01</td>
</tr>
<tr>
<td>spinalis thoracis</td>
<td>-50.5</td>
</tr>
<tr>
<td>longus coli</td>
<td>+13.5</td>
</tr>
<tr>
<td>semispinalis thoracis</td>
<td>0</td>
</tr>
<tr>
<td>spinalis cervicis</td>
<td>0</td>
</tr>
<tr>
<td>semispinalis cervicis</td>
<td>0</td>
</tr>
</tbody>
</table>

For all cases spinalis cervicis and semispinalis cervicis were predicted to be unaffected by the curvature of the postures investigated. Also semispinalis thoracis was only slightly affected in the flex1 posture, and splenius cervicis and longus coli showed relatively small changes in all flexed postures.

The greatest effects were predicted for longissimus thoracis in the flex2 and flex3 postures, thoracic iliocostalis in all four postures, and spinalis thoracis in the erect posture. In the erect posture using the curved spinalis thoracis muscle forces decreased the peak percentage value by 50.5%. Even greater, the thrustline due to a curved representation of thoracic iliocostalis was moved to the negative co-ordinate axis in this posture.

Changes for longus coli in all postures and splenius cervicis in the flexed postures showed an increase in thrustline curve. In contrast all other muscle groups showed flattening of the thrustline curve. Figure 4-47 compares the effects for thoracic iliocostalis and spinalis thoracis in the erect posture and Figure 4-48 compares the effects of these muscles and longissimus thoracis in the flex3 posture.
Figure 4-47: Comparing the effects of straight or curved representation of thoracic iliocostalis and spinalis thoracis muscle forces in the erect posture.
Figure 4-48: Comparing the effects of straight or curved representation of thoracic iliocostalis, spinalis thoracis and longissimus thoracis muscle forces in the flex3 posture.

4.5.1.2 Discussion

4.5.1.2.1 Thoracic Iliocostalis

For thoracic iliocostalis, the predicted thrustlines for the straight representation were included in categories five and six indicating a small effect on the overall thrustline. In contrast the curved representation shows a significant flattening effect on the thrustline. This muscle group would be included in category one for the erect posture, category two for the flex posture, and category seven for the flex2 and flex3 postures for the curved representation. Consequently, these muscles have a major role in flattening the thrustline and supporting the spine under high loading conditions.
For this muscle only the forces applied at the upper attachments are considered within the end limits A and B. The components of force for the straight representation are negative in the perpendicular direction. Directing the muscle around the thoracic curve opposes this, and results in a muscle with a positive, but small perpendicular component of force, and a consequent small decrease in parallel component. This is illustrated in Figure 4-49.

![Diagram of FLEX1 POSTURE showing straight and curved representations of thoracic iliocostalis in the flex1 posture.](image)

Figure 4-49: Lines of action for the straight and curved representation of thoracic iliocostalis in the flex1 posture.

The moments generated by the perpendicular component oppose those due to body weight and result in a small reduction in the end support reactions. However, the most significant effect is with the direct application of the muscle forces on the thrustline.
In the erect posture the small perpendicular components of the deviated muscle force are greater than the small negative components of body weight, and the overall direction of the thrustline due to body weight is reversed. This explains the projection of the thrustline due to the curved representation of the muscle anterior to the reference line in this posture. In the flexed postures, the increase in body weight perpendicular components results in a smaller effect of the muscle forces on the thrustline. The components of muscle force reduce the gradient of the thrustline curve, although the overall thrustline projection is posterior to the reference line dictated by body weight forces.

4.5.1.2.2 Longissimus Thoracis

For the straight representation of longissimus thoracis, the effects of muscle force diminished with flexion and for the flex2 and flex3 postures the thrustline was projected posterior to the reference line. In contrast for the curved representation the predicted thrustline was projected anterior to the reference line in all postures and showed a steeper negative curve with flexion. The effects of curvature were therefore significant. However, for other postures no effect was observed. This can be explained by the method of force representation.

The muscle line of action was assumed to be deviated if the location of the reference markers along the curve between the two points of muscle attachment was greater in the x direction than the points of muscle attachments themselves. For spinalis thoracis, the points of attachment and reference markers were the spinous processes. The position of each spinous process was compared with the positions of the points of muscle attachment to see which point corresponded to the peak of the curve. In each posture, the peak of the curve between these two muscle attachment points was greater, resulting in the muscle directed towards this point.
In the erect and flex1 postures, the attachment of longissimus thoracis fascicles to the transverse and spinous processes resulted in an uneven pair of positions against which to compare spinal curvature. Selection of the spinous processes as reference markers would result in a sudden deviation of the muscle from the transverse process end attachment point outwards towards the nearest spinous process attachment point (Figure 4-50).

Therefore the transverse processes at each level were used as reference markers. Consequently, the peak curvature referenced by transverse processes was not greater than the location of the spinous process, and no deviation resulted.
Although in the erect posture, the longissimus thoracis muscle was not affected by curvature using this representation, even the straight muscle representation results in a flattening of the thrustline far greater than that required to support body weight. The forces generated are therefore ample to support the spine in this posture without a more elaborate method of representation. In the flexed postures in which greater force is required, this representation was more effective as required. Consequently, with the curved representation this muscle was included in category one in all postures.

For longissimus thoracis most of the forces affected by curvature are those which span the thoracic region of the spine and are attached at both ends within the end limits A and B. The perpendicular components of both these forces are increased in the positive direction. In contrast the parallel components of these forces are in opposing directions. The lines of action for both representations are illustrated in Figure 4-51 for the flex3 posture.

![Figure 4-51: Lines of action for the straight and curved representation of longissimus thoracis in the flex3 posture.](image)

The increase in perpendicular component of force in the positive direction generated at both ends of attachment to the spine resulted in a greater amount of force acting against body weight. Consequently, the moments generated by this muscle and the local application of forces opposes body weight and are sufficient to reverse the overall projection of the thrustline.
The thrustline is therefore projected posterior to the reference line in the flex2 and flex3 postures when the muscle line of action is straight, and anterior to the reference line when the muscle line of action is deviated by curvature.

4.5.1.2.3 Spinalis Thoracis, Splenius Cervicis and Longus Coli

For spinalis thoracis the increase in flattening effect observed for the erect posture due to the curved representation would move it from category six to category two. For all other postures, and for longissimus cervicis, a moderate increase in the flattening effects on the thrustline are observed, although the predicted thrustlines remain in these categories.

The effects for splenius cervicis were only slight because the lines of action of this muscle were of shorter span, and were less affected by curvature.

Longus coli which attaches to the anterior of the vertebral bodies in the cervical region was affected by cervical lordosis. The anterior convex curve in the cervical region deviated the muscle force, increasing the negative perpendicular component. Consequently, this component of force acted with body weight, thereby increasing the steepness of the thrustline curve. For this an increase in the predicted thrustline curve both locally and overall were predicted for the erect and flex3 postures in which lordosis was most pronounced. However, these effects were small due to a flatter curvature than that of the thoracic region, and due to the small force magnitude of these muscle fascicles.
4.5.1.3 Summary

The importance of considering the effects of curvature on muscle force in future studies is demonstrated from the results of this study. In flexion when using the straight line representation the effects of longissimus thoracis in flattening the thrustline significantly diminishes, and a combination of several muscle groups maximally contracted would be necessary to support the spine in these postures. Further since these muscles have the greatest effect on the thrustline, with the addition of an external load, the muscle forces may not be sufficient to support the spine. In contrast when the muscle force direction is altered by curvature, a significant increase in its ability to flatten the thrustline and ensure stability occurs. The predicted thrustline for longissimus thoracis lies anterior to the reference line for all four postures, indicating submaximal contraction may be sufficient to ensure stability. Also the muscle force is more likely to be sufficient to support the further application of an external load.

For this reason, the curved representations of longissimus thoracis, and thoracic iliocostalis lumborum which were identified as the major muscles for supporting the spine when flexed, and when subject to external loading, were used in further studies on spinal stability.
4.5.2 VARIATION IN FORCE DISTRIBUTION

4.5.2.1 Inter-segmental Muscles

Little is known about the magnitude or distribution of the cross sectional area of the inter-segmental muscles. It was therefore considered necessary to investigate to what extent a variation in distribution pattern of these forces affects the predicted thrustline.

Due to the attachments of interspinalis and intertransverse at approximately the same levels on the spine, only the interspinalis muscle group was tested. Cervical and thoracic multifidus and rotatores which span different levels were also considered.

The patterns were derived based upon the initial value assumed for the uniform distribution. It was ensured the total applied force was the same in each case so that a comparison between them could be made. An arbitrary increase of 2N at each force level was used to describe a uniform increase or decrease in force descending the spine. A constant force applied in each region, but different between regions was also considered. A description of the loading patterns for each case investigated is presented in Table 4-9. Cases in which the cervical muscle forces were greater than the thoracic or lumbar were justified by a possible requirement for additional force in this region to support the weight of the head. Also, cases in which lumbar and thoracic muscle forces were greater than cervical forces were justified by the need for greater muscle force to support the increase in body weight descending the spine.
**Table 4-9: The force distribution patterns investigated for the intersegmental muscles.**

<table>
<thead>
<tr>
<th>Case Description</th>
<th>Muscle Name</th>
<th>Upper levels</th>
<th>Mid levels</th>
<th>Lower levels</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lumbar&gt;cervical (4060)</td>
<td>Interspinalis</td>
<td>6 pairs at 40N</td>
<td>(2 pairs at 50N)</td>
<td>6 pairs at 60N</td>
</tr>
<tr>
<td>Cervical&gt;lumbar (6040)</td>
<td>Interspinalis</td>
<td>6 pairs at 60N</td>
<td>(2 pairs at 50N)</td>
<td>6 pairs at 40N</td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>Interspinalis</td>
<td>from 37N</td>
<td>-</td>
<td>to 63N</td>
</tr>
<tr>
<td>descending (inc)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>Interspinalis</td>
<td>from 63N</td>
<td>-</td>
<td>to 37N</td>
</tr>
<tr>
<td>ascending (dec)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>rotatores</td>
<td>from 20N</td>
<td>-</td>
<td>to 40N</td>
</tr>
<tr>
<td>descending (inc)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>rotatores</td>
<td>from 40N</td>
<td>-</td>
<td>to 20N</td>
</tr>
<tr>
<td>ascending (dec)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Steps at 2N, greatest</td>
<td>rotatores</td>
<td>from 25N</td>
<td>(2 pairs at 35N)</td>
<td>to 25N</td>
</tr>
<tr>
<td>mid region (mid)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thoracic&gt;cervical (2040)</td>
<td>multifidus</td>
<td>6 pairs at 20N</td>
<td>(5 pairs at 30N)</td>
<td>6 pairs at 40N</td>
</tr>
<tr>
<td>Cervical&gt;thoracic (4020)</td>
<td>multifidus</td>
<td>6 pairs at 40N</td>
<td>(5 pairs at 30N)</td>
<td>6 pairs at 20N</td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>multifidus</td>
<td>from 14N</td>
<td>-</td>
<td>to 46N</td>
</tr>
<tr>
<td>descending (inc)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>multifidus</td>
<td>from 46N</td>
<td>-</td>
<td>to 14N</td>
</tr>
<tr>
<td>ascending (dec)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

**4.5.2.2 Multi-segmental Muscles**

The effects of force distribution was also investigated for larger more superficial muscles. One group from each of the cervical and thoracic regions were studied. The semispinalis capitis and curved representation of spinalis thoracis muscles were chosen as these were considered to have the greatest effect on the thrustline for each region, and would therefore magnify any effects which distribution might have.

Based upon similar loading patterns for rotatores, three loading patterns were considered for each of these muscles. These are summarised in Table 4-10.
Table 4-10: The force distribution patterns investigated for the multisegmental muscles.

<table>
<thead>
<tr>
<th>Case Description</th>
<th>Muscle Name</th>
<th>Upper levels</th>
<th>Mid levels</th>
<th>Lower levels</th>
</tr>
</thead>
<tbody>
<tr>
<td>Increment of 2N</td>
<td>semispinalis capitis</td>
<td>from 46N</td>
<td>(2 pairs at 54N)</td>
<td>to 62N</td>
</tr>
<tr>
<td>descending (inc)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>semispinalis capitis</td>
<td>from 62N</td>
<td>(2 pairs at 54N)</td>
<td>to 46N</td>
</tr>
<tr>
<td>ascending (dec)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Steps at 1N,</td>
<td>semispinalis capitis</td>
<td>from 52N</td>
<td>(2 pairs at 56N)</td>
<td>to 52N</td>
</tr>
<tr>
<td>greatest mid</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>region (mid)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>spinalis thoracis</td>
<td>from 10.125N</td>
<td>-</td>
<td>to 16.125N</td>
</tr>
<tr>
<td>descending (inc)</td>
<td>(curved)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increment of 2N</td>
<td>spinalis thoracis</td>
<td>from 16.125N</td>
<td>-</td>
<td>to 10.125N</td>
</tr>
<tr>
<td>ascending (dec)</td>
<td>(curved)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Steps at 1N,</td>
<td>spinalis thoracis</td>
<td>from 12.125N</td>
<td>(2 pairs at 15.125N)</td>
<td>to 12.125N</td>
</tr>
<tr>
<td>greatest mid</td>
<td>(curved)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>region (mid)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

4.5.2.3 Model Predictions

For each case considered, the greatest flattening effect was observed at the peak of the thrustline curve. The peak x co-ordinate of each thrustline was therefore determined and expressed as a percentage of the peak co-ordinate of the thrustline predicted by body weight alone. This was termed the 'peak percentage' and was used as a comparative measure for the effects of each force distribution.

The force distribution patterns which resulted in the greatest increase and decrease in peak percentage value are reported in Table 4-11. For each value the corresponding case description label is also reported. Positive values indicate an increase in thrustline curve, and negative values a decrease.
Table 4-11: Maximal increase and decrease in peak percentage due to force distribution.

<table>
<thead>
<tr>
<th>Loading Case</th>
<th>Erect %</th>
<th>Flex1 %</th>
<th>Flex2 %</th>
<th>Flex3 %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Interspinalis</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increased thrustline curve</td>
<td>1.6 (dec)</td>
<td>1.91 (6040)</td>
<td>3.99 (6040)</td>
<td>1.2 (6040)</td>
</tr>
<tr>
<td>Decreased thrustline curve</td>
<td>-1.1 (inc)</td>
<td>-1.53 (inc)</td>
<td>1.8 (4060)</td>
<td>-0.8 (4060)</td>
</tr>
<tr>
<td>Rotatores</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increased thrustline curve</td>
<td>0.4 (dec)</td>
<td>0.11 (inc)</td>
<td>0.18 (inc)</td>
<td>0.2 (inc)</td>
</tr>
<tr>
<td>Decreased thrustline curve</td>
<td>-0.3 (inc)</td>
<td>-0.06 (dec)</td>
<td>-0.11 (dec)</td>
<td>-0.1 (dec)</td>
</tr>
<tr>
<td>Cervical and thoracic multifidus</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increased thrustline curve</td>
<td>12.1 (dec)</td>
<td>0.86 (2040)</td>
<td>8.01 (inc)</td>
<td>3.3 (2040)</td>
</tr>
<tr>
<td>Decreased thrustline curve</td>
<td>-14.1 (inc)</td>
<td>-0.75 (4020)</td>
<td>-5.92 (dec)</td>
<td>-2.4 (4020)</td>
</tr>
<tr>
<td>Splnalis thoracis</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increased thrustline curve</td>
<td>0.6 (dec)</td>
<td>0.35 (inc)</td>
<td>1.12 (inc)</td>
<td>1.3 (inc)</td>
</tr>
<tr>
<td>Decreased thrustline curve</td>
<td>-0.7 (mid)</td>
<td>-0.32 (dec)</td>
<td>-1.02 (dec)</td>
<td>-0.1 (mld)</td>
</tr>
<tr>
<td>Semispinalis capitis</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Increased thrustline curve</td>
<td>3.1 (dec)</td>
<td>0.2 (inc)</td>
<td>0.14 (dec)</td>
<td>0.12 (dec)</td>
</tr>
<tr>
<td>Decreased thrustline curve</td>
<td>-2.31 (inc)</td>
<td>0 (mid)</td>
<td>-0.1 (inc)</td>
<td>-0.09 (inc)</td>
</tr>
</tbody>
</table>

The most effective loading pattern varied for each muscle group and also depended upon posture. In terms of increasing the flattening effect on the thrustline, the greatest peak percentage changes were observed for multifidus (14.1%) in the erect posture and 5.92% in the flex2 posture. Considerably smaller values were observed for the remaining muscles, with the greatest value of 2.31% for semispinalis capitis. The amount by which each force pattern differed varied with muscle group and with posture. The predicted thrustlines for the loading patterns in the erect posture for which the effects were greatest are shown in Figure 4-52.
Figure 4-52: Showing the effects of variation in force distribution pattern of the cervical and thoracic multifidus and semispinalis capitis muscles on the predicted thrustline when combined with body weight in the erect posture.

The force pattern which resulted in the greatest flattening effect on the thrustline for each muscle group and posture is as follows:

**Interspinalis:**
- greatest force applied in lumbar region for all postures

**Rotatores:**
- greatest force at the upper thoracic levels for the erect posture
- greatest force at the lower levels in the flexed postures

**Semispinalis capitis:**
- greatest force at the upper levels for erect, flex2 and flex3 postures
- uniform and mid distributions same in flex1 posture
Cervical and thoracic multifidus:
greatest force in the thoracic region for erect posture
greatest force in the cervical region for flexed postures

Spinalis thoracis:
greatest force mid region in the erect and flex3 posture
greatest force at the lower levels in the flex1 and flex2 postures

4.5.2.4 Discussion

Initially loading patterns for interspinalis and multifidus distinguished between a gradual change in force at each vertebral level (dec or inc) and the application of forces of the same magnitude within a region, but of different magnitude between regions (cases 4060, 6040, 2040 and 4020). However, due to the small variation predicted for each loading pattern, this detail was revealed to be unnecessary. Indeed for all but the multifidus muscle, a variation of less than 2.5% was observed.

For each muscle group, the effectiveness of the loading pattern depends upon the direction of applied forces at each level. The variation in force direction with posture explains the variation in loading pattern observed.

For multifidus, in the erect posture forces at the upper cervical levels have a greater negative perpendicular component of force which increase the overall thrustline curve relative to body weight. Application of the greater force at lower (thoracic) levels therefore reduces this effect on the thrustline.
In the flexed postures, the relative positions of the points of force application change in the cervical region, and the force direction is altered. These forces have positive perpendicular components of force, which increase the flattening effect on the spine. Application of greater force in this region therefore enhances this effect.

For the interspinalis muscles in the erect posture the direction of forces follows both cervical and lumbar lordotic curves. Application of the larger forces at the lumbar levels allows the flattening effect of these muscle forces to take effect sooner on the predicted thrustline curve.

In the flexed postures, the convex curve of the spine is increased, and forces applied in the lumbar regions are directed posteriorly with positive perpendicular components of force. In the cervical region, this direction is reversed. A force distribution pattern which favours the greater forces with positive perpendicular components increases the flattening effect. Hence the force distribution pattern with the greatest force in the lumbar region showed the greatest flattening effect.

The greatest flattening effect for semispinalis capitis was predicted with the force distribution pattern which decreased in magnitude descending the spine. At the upper levels of the spine, the forces are directed with positive perpendicular and parallel components, which oppose the direction of forces due to body weight. The force distribution which applies the greatest force at these upper levels therefore results in the greatest flattening effect.

The greatest difference was observed between patterns in the erect posture. In the flexed postures, due to an increase in thrustline curvature caused by the increased perpendicular direction of body weight forces, the effects of the muscle forces are reduced, and the difference between them is smaller.
For rotatores and spinalis thoracis, smaller changes were observed with force distribution due to very small differences between the direction of the fascicles.

Due to variation in total muscle cross sectional area and the number of levels spanned, the arbitrary increase or decrease of 2N at each level represents a different variation in force relative to the initial value. For example for the interspinalis, a range of 40 to 60N represented a variation force of 20% above and below the initial value of 50N. In contrast for semispinalis capitis a range of 46 to 64 N represented a variation force of 14.8% above and below the initial value of 54N. The precise force distribution patterns are unknown. However, based upon the predictions, the effects of assuming a uniform force distribution are small.

For all cases except semispinalis capitis in the flex1 posture, the uniform force pattern was within the range of predicted effects, and was therefore considered to represent the 'mean' behaviour of the thrustlines. Also due to the very small effects which occurred with a variation in loading distribution, the uniform force distribution pattern was considered to be a good way to approximate the unknown force pattern of these muscles and was therefore used in further study. However, when considering combinations of muscle forces which ensure stability a small increase in muscle force may distinguish between whether or not conditions of stability are satisfied. This is considered in later investigations on stability (p320 and 326)
4.5.3 THE EFFECTS OF FORCE MAGNITUDE ON THE PREDICTED THRUSTLINE FOR THE INTERSEGMENTAL MUSCLES

The close proximity of the intersegmental muscles to the spine makes their activity difficult to monitor using EMG electrodes. Their precise role in various postures is therefore unclear. Also the pathway of more superficial muscles obscures the intersegmental muscles in cross sectional studies, thereby making it difficult to determine their cross sectional area and to estimate the amount of force they can generate. The effects of an increase in force for these muscles was therefore investigated.

In this study one cross sectional area of 50 mm$^2$ for the interspinalis and 25 mm$^2$ for the intertransverse muscles between the centre and upper torso were reported (Frievalds, 1985). In the lumbar regions of the spine, the values for these muscles may be greater. The points of attachment of both muscle groups in the sagittal plane are similar. Therefore the interspinalis muscle was chosen as a representative muscle for both. Assuming the cross sectional area of this muscle was doubled, the predicted thrustlines were calculated.

For thoracic rotatores, data in the literature could not be found. Assuming this muscle to be small, it was assigned a value of 30N for each fascicle. To investigate the effects of a larger muscle group, the force this muscle could generate was doubled, and the thrustlines predicted.
For cervical multifidus, values reported in the literature varied considerably. In my own study of cross sectional anatomy drawings, the scaled area observed was 0.35 cm\(^2\). However, this muscle was visible for very few levels in these drawings, so that the value obtained may not be the maximum. Frievalds (1985) reported a larger value of 1.25 cm\(^2\) also using anatomy drawings of cadaver specimens whereas Mayoux Benhamoux et al. (1989) reported a value of 5.80 cm\(^2\) at the C5 level. This latter value is greatest for the values reported for all cervical muscles, implying that cervical multifidus is the strongest neck muscle. However, such a large muscle was not apparent in the cross sectional anatomy drawings. Possibly the boundaries of muscle may have been combined with those of another in the CT scans. To cover a realistic range for the magnitude of force generated by this muscle, a value of 100N was assigned to each fascicle supplementing the existing study of 30N.

4.5.3.1 Model Predictions

The ‘peak percentage’ defined in the previous investigation for force distribution pattern (page 322) was used as a comparative measure for the overall flattening effects of the muscle forces on the predicted thrustline.

4.5.3.1.1 Erect Posture

Figure 4-53 compares the predicted thrustlines for the initial nominal force values and the defined increase. The peak percentage value was 2.63% less with a doubling of force for interspinalis, and 0.24% less with a doubling of force for the thoracic rotatores. The approximate shape of the thrustlines was similar. Doubling the magnitude of these forces therefore had a very small effect on the predicted thrustline in this posture. In contrast a three-fold increase in the force of multifidus resulted in a 36.9% increase in the peak percentage value, resulting in a significant increase in thrustline curvature.
4.5.3.1.2 Flexed Postures

With flexion the difference in peak percentage values with an increase in magnitude was again very small for rotatores, but was more significant for interspinalis and multifidus. The peak percentage values are summarised in Table 4-12. The greatest effect for interspinalis was in the flex2 posture (9.93%).

Table 4-12: Decrease in the peak percentage value with an increase in force magnitude for the intersegmental muscles.

<table>
<thead>
<tr>
<th>Posture</th>
<th>Interspinalis %</th>
<th>Cervical and Thoracic Multifidus %</th>
<th>Rotatores %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect</td>
<td>2.63</td>
<td>36.9</td>
<td>0.24</td>
</tr>
<tr>
<td>Flex1</td>
<td>8.28</td>
<td>7.5</td>
<td>0.3</td>
</tr>
<tr>
<td>Flex2</td>
<td>9.93</td>
<td>26.1</td>
<td>0.71</td>
</tr>
<tr>
<td>Flex3</td>
<td>7.94</td>
<td>10.4</td>
<td>0.6</td>
</tr>
</tbody>
</table>

Figure 4-53: Showing the effects on the predicted thrustlines with an increase in force magnitude in the erect posture.
Figure 4-54 shows the changes in predicted thrustlines for the flex3 posture. For the multifidus muscle, an abrupt change in thrustline path occurs at the upper levels of the spine. This is due to the change in direction of forces at the upper levels, and is magnified with the increase in muscle force.

Figure 4-54: Showing the effects on the predicted thrustlines with an increase in force magnitude in the Flex3 Posture.
4.5.3.2 Discussion

The increase in force magnitude investigated were selected to represent an upper bound for the anatomic cross sectional areas of the inter-segmental muscles. In the lumbar regions, the individual muscle forces of the longissimus thoracis, which is one of the largest muscle groups attaching to the spine, were applied based on cross sectional areas ranging from 29 to 178 mm². Although the cross sectional area of the inter-segmental muscles at each level has not yet been reported, the closer attachment to the vertebral body centres, and the short span of their fascicles indicates a smaller cross sectional area. Upper bounds of 60 mm² for the rotatores, 100 mm² for the multifidus and interspinalis were therefore considered reasonable. The predicted thrustlines determined therefore represented the greatest effects likely for each muscle force for realistic anatomical limits. These effects were considered in later investigations (p350 and 356).
4.6 MUSCLE GROUPS WHICH ARE IMPORTANT FOR STABILITY OF THE SPINE

The study of the independent effects of the muscle and ligament forces when combined with body weight showed no single muscle force set satisfied conditions of stability by itself. Combinations of muscle forces, are therefore needed for stability of the spine.

Based upon the understanding for each individual muscle and ligament group, the activity of these groups could be matched to the force requirements for each posture. Combinations of muscle and ligament forces could then be selected and used to determine the stability of the spine in each posture when supporting body weight and external loads.

4.6.1 MATCHING MUSCLE AND LIGAMENT BEHAVIOUR WITH THE FORCE REQUIREMENTS IN EACH POSTURE

In the erect posture, the forces due to body weight generate a thrustline which follows a posterior convex curve. The overall depth of this curve is greater than the convex curve of the thoracic region. Some muscle and ligament force is therefore needed to reduce the overall depth of the curve. Also the path of the thrustline in the cervical and lumbar regions must be altered significantly to follow the anterior convex curves in these regions.
Muscle forces applied in the cervical and lumbar regions were observed to result in localised flattening of the thrustline in these regions. These also resulted in an overall reduction in the depth of the thrustline curve moving the thrustline closer to the vertebral configuration in the thoracic region. To prevent over-flattening of the thrustline curve in the thoracic region muscles such as iliocostalis cervicis and scalenus posterior may also be necessary in the thoracic region.

Also the interspinalis and intertransverse muscles and the ligaments which generate force along the path of the spine were considered for their contribution to stability in this posture.

With flexion the spine moves towards a configuration which follows a posteriorly convex curve, although the predicted thrustline due to body weight alone is much steeper. Greater muscle and ligament force is therefore required to reduce the depth of the overall thrustline. Muscles with larger effects on the overall thrustline curve were therefore considered.

Also, to investigate the flexion-relaxation phenomenon (e.g. Schultz et al. 1985) the contribution of the intersegmental muscles and ligaments towards stability in this posture were considered.
4.6.2 MUSCLE GROUPS REQUIRED TO STABILISE THE SPINE WHEN SUPPORTING THE WEIGHT OF THE UPPER BODY

Matching muscle and ligament groups to the stability requirements of the spine in each posture indicates possible combinations which could ensure stability. However, for each combination a unique set of forces, generated moments and end support reactions affects the resulting thrustline. The resulting thrustline is therefore not simply the superimposition of the predicted thrustlines for each individual muscle group. Quantitative testing of the possible combinations of muscles and ligaments was therefore required.

Based upon the predicted thrustline for a given combination, additional muscle forces were added, or muscle forces removed, or reduced (representing submaximal contraction) until a close match between thrustline and vertebral configuration had been obtained.
4.6.2.1 Erect Posture

Initially only two sets of forces for this posture were considered. Intra abdominal pressure (4KPa) was identified to have significant localised flattening effect on the thrustline in the lumbar region. This was combined with selected cases of cervical muscles:

- longissimus capitis
- longus capitis
- spinalis capitis
- intertransverse

Of these, the combination of intra abdominal pressure and longissimus capitis produced the closest fit thrustline (shown in Figure 4-55).

Assuming a lower value of intra abdominal pressure of 1KPa, typical of relaxed standing (Schultz et al., 1982), additional force was required in the lumbar region. The lumbar multifidus submaximally contracted to 40%, together with the intertransverse muscle was found to provide a close fit although slight instability occurred in the cervical and lumbar regions.
For this latter combination the predicted thrustline was close to the anterior border of the thoracic vertebral bodies so that further force in the cervical and lumbar regions would move the thrustline beyond this limit in the thoracic region. However, addition of the iliocostalis cervicis muscle opposed this, moving the thrustline posterior at the upper vertebral levels. A greater contraction of multifidus (70%) and additional force from the interspinalis muscle was then required, although the thrustline was close to the vertebral body posterior border at the upper thoracic levels and was still outside the vertebral bodies in the lumbar regions. A fine balance of muscle force to ensure stability in this posture is therefore difficult. These thrustlines are shown in Figure 4-55.

Figure 4-55: Thrustlines for the erect posture which closely fit the outlines of the vertebral bodies.
The combined effects of the intersegmental muscles intertransverse and interspinalis were found to result in localised flattening of the thrustline in the cervical and lumbar regions, although this was insufficient to ensure stability. The addition of the ligamentum flavum enhanced this effect by a small amount (Figure 4-56). In contrast, the addition of intra abdominal pressure (4KPa) to these combinations was found to move the thrustline much closer to the spine.

The sensitivity analysis performed for the intersegmental muscles predicted the greatest effect on the peak thrustline co-ordinate to be 1.6% with a variation in force distribution pattern and 2.63% increase with a doubling of force magnitude for the interspinalis muscle. The effects of an increase in force magnitude were therefore included. However, only a very small further effect was predicted (Figure 4-56).

Figure 4-56: Thrustlines for the erect posture due to combinations of intersegmental muscles, ligaments and intra abdominal pressure.
4.6.2.2 Flex1 Posture

In the flex1 posture the shape of the predicted thrustline produced by semispinalis capitis followed the convex curves of the cervical and thoracic regions, although maximal contraction resulted in an over-flattening of the thrustline. Also contraction of this muscle resulted in a thrustline which is opposite in curvature to the anteriorly convex lumbar curve. Some forces in the lumbar region are therefore required to ensure stability in this region.

Figure 4-57 shows thrustlines for combinations of semispinalis capitis with lumbar muscles which resulted in close fit thrustlines. Combinations of 40% semispinalis capitis with maximal contraction of either lumbar iliocostalis or multifidus resulted in a flattening of the thrustline slightly greater than that required in the thoracic regions. Alternatively the combination of intra abdominal pressure (6KPa) and submaximal contraction of semispinalis capitis (25%) and lumbar iliocostalis (55%) provided a thrustline close to the centre line of the spine. Stability of the spine at all levels was thus obtained with this combination.
Figure 4-57: Thrustlines for the flex1 posture which closely fit the outlines of the vertebral bodies.
4.6.2.3 Flex2 Posture

In the flex2 posture, the semispinalis capitis muscle produces a thrustline which fits the path of the vertebral bodies in the cervical and thoracic regions, but which is slightly too steep in the lumbar regions. Addition of either lumbar multifidus or lumbar iliocostalis which flatten the thrustline in the lumbar region resulted in thrustlines coinciding with the path vertebral bodies as required for stability. Submaximal contraction (to various degrees) of semispinalis capitis and longissimus capitis also resulted in close fitting thrustlines. These are illustrated in (Figure 4-58).

![Figure 4-58: Thrustlines for the flex2 posture which closely fit the outlines of the vertebral bodies.](image-url)
4.6.2.4 Flex3 Posture

In the flex3 posture, maximal contraction of the curved representation of longissimus thoracis muscle results in excess flattening of the thrustline. This was found to be the only muscle which could independently flatten the thrustline beyond the vertebral body limits. Stability could be ensured by sub-maximal contraction of this muscle.

The semispinalis capitis muscle was also found to result in considerable flattening of the predicted thrustline due to body weight, although this was insufficient to fit the path of the vertebral bodies. Combinations of this muscle with other muscles were tested:
iliocostalis lumborum
intra abdominal pressure
lumbar multifidus
splenius capitis.
These showed a flattening of the thrustline in the lumbar region but did not ensure stability in the cervical and thoracic regions (Figure 4-59).

Triple combinations of semispinalis capitis, splenius capitis and either intra abdominal pressure or iliocostalis lumborum had little further effect. Activation of the longissimus thoracis muscle was therefore needed.
Figure 4-59: Predicted thrustlines for semispinalis capitis combined with lumbar muscle groups in the flex3 posture.

A combination of longissimus thoracis (curved) submaximally contracted at 25% and semispinalis capitis was found to result in a thrustline which closely fitted the path of the vertebral bodies. Very slight instability occurred in the upper thoracic and lumbar and cervical regions (Figure 4-60).
Also for this posture to test the capacity of the deep muscles and ligaments of the back associated with the flexion-relaxation phenomenon, a combination of interspinalis, intertransverse and ligamentum flavum and cervical and thoracic multifidus was also considered. Figure 4-61 shows the thrustline due to body weight is flattened significantly under the action of this combination of muscle and ligament forces, although this is insufficient to ensure stability at any spinal level.

Results of the sensitivity analysis indicated greater changes in thrustline with magnitude than with force distribution. Changes in peak percentage for interspinalis and multifidus were 1.2% and 2.4% with force distribution and 7.94% and 10.4% with magnitude. An increase in force magnitude for the intersegmental (including intertransverse) muscles was therefore tested. A further flattening effect was predicted, although again this was insufficient to ensure stability at any level of the spine.
Figure 4-61: The effects of the intersegmental muscles and ligamentum flavum on the predicted thrustlines due to body weight for the flex3 posture. (multifidus = cervical and thoracic)

4.6.2.5 Discussion

For each posture, thrustlines are predicted which fit closely with the path of the vertebral bodies.

At levels where the thrustline passes slightly posterior to the vertebral bodies, the facets may supplement the load bearing role of the intervertebral discs. This may be significant in the erect posture for which the cervical and lumbar vertebrae are extended.
In the erect posture combinations of two maximal force groups (intra abdominal pressure and longissimus capitis) were sufficient to ensure stability. Alternatively combinations of sub-maximal force groups (e.g. lumbar multifidus at 20%, intra abdominal pressure at 4KPa and inter transverse at 100%) were tested and were also shown to provide a close fit between thrustline and vertebral configuration. Similarly, in the flexed postures, several combinations of maximal and submaximal force groups were tested.

Numerous other muscle groups not included indicates the available muscle force to support the spine is in excess of that required. Contraction of these forces to provide fine adjustments to the path of the thrustline undoubtedly occurs.

A number of thrustlines can be produced which satisfy stability according to the muscle recruitment pattern. Biological variation in the dimensions of the spine, the stiffness properties, the range of configurations and the muscle strengths, means the precise combination of muscle forces which ensures stability for a given posture for a given individual is unique. However, the existence of at least one thrustline for the given posture indicates stability for the spine.

The main finding of this study is that the muscle groups tested in this study are capable of ensuring stability of the spine and supporting the weight of the upper body for the postures tested and that the ligaments have a comparatively small contribution.
4.6.3 MUSCLE GROUPS REQUIRED TO STABILISE THE SPINE WHEN SUPPORTING THE WEIGHT OF THE UPPER BODY AND AN EXTERNAL LOAD OF 900N

The amount of muscle force required in each posture was increased with the addition of an external load. For the erect posture this increase was smallest due to the large parallel component of the external load contributing to the compressive thrust required. In contrast, with flexion, the significant increase in thrustline curvature indicated a greater force requirement.

4.6.3.1 Erect Posture

With an increase in the curve of the thrustline due to the external load, it was estimated slightly more force was required in the cervical and lumbar regions to control the path of the thrustline. However, a precise combination of forces which satisfied stability was difficult to determine.
Sufficient force could not be generated to satisfy stability in the lumbar regions without moving the overall thrustline anterior to the reference line. The addition of muscles in the thoracic regions which opposed this effect and would increase the thrustline curve so that it corresponded to the thoracic curve of the spine were therefore considered. These included iliocostalis cervicis, scalenus posterior and submaximal activation of rectus abdominis. The effects of including the thoracic muscle iliocostalis cervicis can be appreciated in Figure 4-62. A combination of longissimus capitis (50%) and intra abdominal pressure (8KPa) did not provide sufficient force to ensure stability in the cervical or lumbar levels but additional force in these regions would result in the thrustline moving anterior to the spine in the thoracic regions. Addition of the iliocostalis cervicis muscle opposed this effect in the thoracic region allowing more muscle force to be applied in the cervical and lumbar regions (IAP (12KPa) and intertransverse).
The above combination of longissimus capitis, IAP (8KPa), iliocostalis cervicis and intertransverse resulted in the best fit between thrustline and vertebral curve. A series of combinations was investigated using submaximal activity of rectus abdominis, although the precise local control of the thrustline could not be achieved without significantly altering the balance of moments and the overall thrustline position.
4.6.3.2 Flex1 Posture

Initially, the effects of the longissimus thoracis (curved representation) in supporting the external load were investigated. The predicted thrustline was approximately mid-way between the predicted thrustline for the external load and the configuration of the spine. This muscle therefore supplies approximately half the required force. The addition of the thoracic part of iliocostalis lumborum showed a further flattening of the thrustline and the further addition of semispinalis capitis resulted in a thrustline very close to the path of the vertebral bodies.

Further combinations which showed a close fit to the spine configuration were

ii) longissimus thoracis, semispinalis capitis, intra abdominal pressure and iliocostalis lumborum

iii) longissimus thoracis, semispinalis capitis, intra abdominal pressure and lumbar multifidus

These are shown in Figure 4-63. Sufficient force is therefore available to support body weight and an external load in this posture.
4.6.3.3 Flex2 and Flex3 Posture

In the flex2 posture, the combination of longissimus thoracis (curved) and thoracic iliocostalis (curved) and semispinalis moved the thrustline close to the vertebral path although instability occurred at the upper thoracic regions. The addition of splenius capitis had only a small further flattening effect. The predicted thrustlines are shown in Figure 4-64.
Figure 4-64: Thrustlines for muscle combinations which closely fit the outlines of the vertebral bodies acting with body weight and an external load of 900N in the flex2 posture.

Similarly for the flex3 posture, for these muscle combinations instability was present in the upper thoracic and the cervical regions of the spine and was greater.
A number of alternative muscle combinations were considered:

i) The addition of remaining cervical muscles of category one for this posture: splenius cervicis and longissimus capitis.

ii) An increase in the force applied at the lumbar regions by the addition of lumbar iliocostalis and multifidus and intra abdominal pressure

iii) A 10% increase in all force (semispinalis capitis, longissimus thoracis and thoracic iliocostalis) to account for an increase in muscle area for living tissue (rather than values estimated from cadaver tissue).

iv) Additional forces due to interspinalis and intertransverse included to represent the effect of submaximal activity of other muscle groups.

The predicted thrustlines for these cases are shown in Figure 4-65 for the flex3 posture. Each case had a small effect on the predicted thrustline. The greatest was for the addition of force in the lumbar region due to iliocostalis, multifidus and intra abdominal pressure. However, this had greatest effect on the lumbar regions rather than the thoracic regions and resulted in further instability at the lumbar levels. For all cases, the peak of the thrustline remained. Similar results were observed for the flex2 posture. The results of this study for the flex2 and flex3 postures therefore indicated insufficient force to be applied in the thoracic region when supporting an external load of 900N.
4.6.3.4 Extended Study for the Flex2 and Flex3 Postures

It was hypothesised that the additional force required in the thoracic regions to support the external load could be provided by the shoulder muscles. Attachment of these muscles to the spine ranges from C1-C4 for the levator scapula, C7 to T4 for the rhomboids, T1 to T12 for the trapezius and T6-T12 for latissimus dorsi. The forces generated by these muscles to stabilise the shoulder complex help to support the external load carried and are distributed over the entire length of the thoracic spine.
To test this forces representing the trapezius muscle were added to the existing muscle combination of longissimus thoracis, thoracic iliocostalis and semispinalis capitis. Data regarding the points of attachment and cross sectional areas were obtained from the study of Johnson et al. (1994). In comparison with the predicted thrustline for the extensor muscle combination, the addition of the trapezius muscle moves the thrustline significantly closer to towards the path of the vertebral bodies in the thoracic region (Figure 4-66).

For the flex3 posture, a similar but smaller effect occurred. Additional force due to latissimus dorsi was therefore considered. The cross sectional anatomy drawings were used to identify a representative point of attachment on the humerus approximately 3.5 cm anterior to the T3 vertebra. Anatomical textbooks revealed this muscle attached to the spinous processes of vertebrae T6-T12 and attached in the lumbar regions to the thoraco lumbar fascia. To approximate the action of this muscle forces acting between the humerus and the spinous processes of vertebrae L5 to T6 were applied. The maximum cross sectional area reported by McGill et al. (1993) of 2596 mm² was assumed to be uniformly distributed among the fascicles.

The addition of this muscle activated to 50% to the combination of extensor muscles and trapezius resulted in a thrustline which followed the path of the vertebral bodies in the flex2 posture (Figure 4-66).
For the flex3 posture, the same combination of longissimus thoracis, thoracic iliocostalis, semispinalis capitis, trapezius and latissimus dorsi activated 100% moved the predicted thrustline closer to the path of the vertebral bodies. Slight instability still occurred at these levels (Figure 4-67). Also, sudden changes in the thrustline at the T3 and T1 levels occur due to the application of large forces at a single point on the humerus and scapula. However, addition of other muscle groups such as levator scapula, rhomboids and serratus anterior will reduce this abrupt change in thrustline direction and will contribute further to the spinal stability. When supporting an external load, the shoulder muscles therefore play a vital role.
Figure 4-67: The effects of shoulder muscle forces on the predicted thrustline due to external load, body weight and extensor muscle forces in the flex3 posture.
4.6.3.5 Discussion

In the erect posture the addition of an external load to the existing body weight pattern did not significantly increase the steepness of the thrustline curve and the amount of muscle force required for support. However, while many possible combinations were investigated, the action of the muscle groups was found to contradict stability of the lordotic cervical and lumbar spine and the kyphotic thoracic spine. The addition of the iliocostalis cervicis muscle which increased the thrustline curve in the thoracic region was necessary to ensure sufficient control. However, slight instability in the lumbar region was still obtained. This may indicate loading shared between the intervertebral disc and facets in this posture for these loading conditions. In vivo, the spine may readjust the spinal configuration if the loads on the facets become too great.

The addition of an external load to the spine in the flexed postures significantly increases the amount of muscle force required for stability. Combinations of muscles which provide a significant flattening of the thrustline in these postures are therefore required. These muscles are longissimus thoracis, thoracic iliocostalis and semispinalis capitis which are considered the main extensors of the spine.

In the flex2 and flex3 postures, maximal contraction of the main extensor forces is not sufficient to ensure stability. In the flex2 posture, the addition of splenius capitis to the combination of longissimus thoracis, thoracic iliocostalis lumborum and semispinalis capitis showed a very small effect on the thrustline. The addition of further cervical muscle groups also has a very small effect. This was also observed for the flex3 posture.
The addition of force sets representing shoulder muscles to these combinations resulted in a significant flattening effect on the thrustline in these postures. The shoulder muscles are therefore required in addition to the spinal extensor muscles to stabilise the spine when supporting an external load.
CHAPTER FIVE:

5. MODEL EVALUATION AND VALIDATION

Using the developed mathematical model, the stability of the spine has been investigated under the action of body weight, an external load and with the individual muscle and ligament groups. Combinations of muscle forces which satisfy stability have also been identified. To evaluate the effects of the greater anatomic detail in this model, a comparison is made with collective loading for each posture. The two types of loading are compared in terms of the stability and joint reactions. These are considered for the effects of body weight loading patterns alone and for the combinations of muscle groups which satisfied stability.

The joint reactions and muscle recruitment pattern predictions are also compared with predictions of the lever model and with in vivo intra discal pressure measurements and EMG data.

Possible developments of this model are then considered.
5.1 EVALUATION OF THE MODEL FEATURES

5.1.1 EFFECTS OF DISTRIBUTED FORCES DUE TO BODY WEIGHT ON THRUSTLINE CURVE

Determination of a thrustline for a given configuration and for a given set of loading conditions is an immediate visible indicator of equilibrium and stability of the spine. A resulting thrustline which matches the path of the vertebral bodies at every level along the entire length of the spine ensures whole spine stability. Application of body weight forces at each vertebral level was therefore considered necessary.

A comparison between the predicted thrustlines for body weight applied using a collective loading pattern and the distributed loading pattern in the model has been made.

For each case the weight of the head (34.59N) was applied at the upper end of the arch in order to provide the necessary compression. For the collective loading case the weight acting at upper vertebral levels C1-T2 (73.36N) was applied at T2 and the remaining weight acting at levels T3-L5 was applied at T12 (272N). For the distributed loading case forces were applied representing body weight at each vertebral level.
The predicted thrustlines due to the distributed and collective loading patterns for body weight are compared in Figure 5-1 for the erect posture and in Figure 5-2 for the flex1 posture. In these figures the outline of the vertebral bodies is also displayed. For the collective loading pattern the predicted thrustline consists of three linear portions which show abrupt changes in direction upon application of force. In contrast the distributed loading pattern results in a thrustline with a greater number of smaller linear portions so that the overall path can be approximated by a smooth curve.

In Figure 5-2 the actual path of the predicted thrustlines for the two loading patterns shows close agreement for the flex1 posture whereas Figure 5-1 for the erect posture a significant difference between the depth of the thrustline curves is observed.

In the erect posture for each loading pattern the total applied force is the same and the end support reaction components H1 and H2 in the parallel direction are identical. However, for each loading pattern, the difference in points of force application results in different generated moments and different values of the end support reaction components R1 and R2 in the perpendicular direction.
For both loading patterns in the erect posture the parallel components of forces due to body weight generate the dominant moments. In the collective loading pattern, the forces due to body weight are applied at the centre of T2 and T12 vertebrae which lie close to the reference line of the spinal arch. In contrast for the distributed loading pattern the forces due to body weight are applied eccentric to the vertebrae and the moment arm of the parallel force component is greater. The parallel components for the distributed loading pattern therefore generate greater moments than those of the collective loading pattern and the end support reaction components R1 and R2 are different.
Table 5-1 shows the moments due to perpendicular components (Fr) of force are similar for both loading patterns in the erect posture (2122.14N and 2368.31N), whereas the moments due to parallel components (Fh) are significantly different (3229.59N and 9697.85N). Due to the eccentricity of the body weight forces in the distributed loading pattern the generated moments are approximately three times as large as those due to the collective loading. Since the directions of these moments for both loading patterns are the same, the increase in moment increases the end support reaction components in the same direction. The initial projection of the thrustline curve for the distributed loading pattern is more inclined to the reference line than that due to the collective loading pattern.

Table 5-1: Comparison of moments generated by the components of body weight force and the resulting end support reaction components R1 and R2 for the collective and distributed loading patterns in the erect and flex1 postures.

<table>
<thead>
<tr>
<th>Loading Pattern</th>
<th>Fr Moments about A (Nmm)</th>
<th>Fr Moments about B (Nmm)</th>
<th>Fh Moments about A/B (Nmm)</th>
<th>R1 (N)</th>
<th>R2 (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect Posture</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Collective</td>
<td>2112.14</td>
<td>-2290.42</td>
<td>3229.59</td>
<td>-1.75</td>
<td>9.98</td>
</tr>
<tr>
<td>Distributed</td>
<td>2368.31</td>
<td>-2034.21</td>
<td>9697.85</td>
<td>-14.32</td>
<td>22.54</td>
</tr>
<tr>
<td>Flex1 Posture</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Collective</td>
<td>3.79E4</td>
<td>-4.04E4</td>
<td>-3582.36</td>
<td>82.27</td>
<td>64.28</td>
</tr>
<tr>
<td>Distributed</td>
<td>4.27E4</td>
<td>-3.56E4</td>
<td>2282.62</td>
<td>84.24</td>
<td>62.32</td>
</tr>
</tbody>
</table>

In the flex1 posture, the increase in perpendicular component of force results in greater moments generated by these components for both loading patterns. The proportional decrease in parallel component results in much smaller moments generated by these components and the difference in overall moments for the two loading patterns is diminished. Consequently the values for R1 and R2 are similar for both loading patterns and the overall depth of the thrustline curves is similar.
Figure 5-2: Comparison for the predicted thrustlines determined for the collective and distributed loading patterns for body weight in the flex1 posture.

With the addition of a 900N load at T2, T3 and T4 for the distributed loading pattern the predicted thrustline (Figure 5-3) is greater than that due to body weight only (Figure 5-1). In contrast for the collective loading pattern the depth of thrustline curve is decreased with the addition of an external load (Figure 5-3).
Figure 5-3: Comparison for the predicted thrustlines determined for the collective and distributed loading patterns for body weight and external load in the erect posture.

Again this is due to the difference in points of force application. Table 5-2 compares the moments and the consequent end support reaction components R1 and R2 due to body weight and an external load applied using the distributed and collective loading patterns in the erect and flex1 postures.
In the erect posture for both loading patterns, the points of application of the external load are posterior to the reference line. The moments due to parallel components of force therefore oppose those due to body weight. However, the point of application at T6 vertebral body centre for the collective loading is further posterior than the points of application at T2, T3 and T4 for the distributed loading. The parallel component of the collective loading force therefore generates the greater moment. This opposes the moment generated by body weight and makes the overall moment due to parallel force negative (Table 5-2). In contrast for the distributed loading the moment due to parallel components of body weight forces is greater and the overall moment due to parallel force is positive (Table 5-2). As a result, although R1 and R2 are of approximately equal total for each loading pattern, R1 is significantly greater than R2 for the collective loading pattern, whereas this is reversed for the distributed loading pattern.

Table 5-2: Comparison of moments generated by the components of body weight force and external loading and the resulting end support reaction components R1 and R2 for the collective and distributed loading patterns in the erect and flex1 postures.

<table>
<thead>
<tr>
<th>Loading Pattern</th>
<th>Fr Moments about A (Nmm)</th>
<th>Fr Moments about B (Nmm)</th>
<th>Fh Moments about A/B (Nmm)</th>
<th>R1 (N)</th>
<th>R2 (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect Posture</td>
<td>8372.60</td>
<td>-6458.06</td>
<td>-7701.41</td>
<td>26.45</td>
<td>1.25</td>
</tr>
<tr>
<td>Distributed</td>
<td>9807.37</td>
<td>-5023.26</td>
<td>4297.64</td>
<td>136</td>
<td>26.35</td>
</tr>
<tr>
<td>Flex1 Posture</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Collective</td>
<td>1.50E5</td>
<td>-1.14E5</td>
<td>-3.77E4</td>
<td>283.76</td>
<td>209.96</td>
</tr>
<tr>
<td>Distributed</td>
<td>1.76E5</td>
<td>-8.82E4</td>
<td>-2.54E4</td>
<td>212.67</td>
<td>281.05</td>
</tr>
</tbody>
</table>
The effects of the different proportions of R1 and R2 result in different thrustline curves which can be explained by considering the proportions of the perpendicular and parallel components of the compression force vector. At each point of force application, negative perpendicular and parallel components of force are added.

Due to the small value for R1 in the distributed loading case, the addition of body weight forces with negative perpendicular components results in a negatively increasing perpendicular component of the compression force vector. In contrast due to the large positive value for R1 for the collective loading pattern, the addition of small negative perpendicular components results in a positive decreasing value. The consequence is a negatively increasing gradient for portions of thrustline for the distributed loading and a positively decreasing gradient for the collective loading case. The increasing gradient for the distributed loading results in a greater thrustline curve (Figure 5-3).

Again in the flexed postures as the moments due to perpendicular components of force increase, the difference in moments due to the parallel components becomes less significant. Consequently the difference between the thrustline curves depth is smaller. This is illustrated in Figure 5-4 for the flex1 posture.

The representation of a smooth thrustline curve obtained with the distributed loading pattern is also diminished as the slope of the thrustline increases and decreases about the peak of the curve at the thoracic levels. and with the addition of muscle forces, flattening of this curve may display a smoother more gradual curve than the collective loading pattern.
Figure 5-4: Comparison for the predicted thrustlines determined for the collective and distributed loading patterns for body weight and external load in the flex1 posture.

5.1.1.1 Conclusions

In all four postures the predicted thrustline for the distributed loading pattern due to body weight resulted in a more gradual change in compression force vector direction which could be approximated by a smooth curve. This is considered to be closer to the gradual curvature of the spine and is a more favourable representation than the few straight line segments predicted for the collective loading pattern.
In the erect posture, the eccentricity of the force due to body weight and external load results in a significant difference in the predicted moments and the consequent depth of the predicted thrustline curve. In particular, the collective loading pattern which applies the body weight forces at the centre of the vertebral bodies and the external load at the T6 vertebra results in a flatter thrustline curve. The collective loading pattern thus underestimates the amount of muscle and ligament force required for stability in this posture.

In the flexed postures, the significance of the eccentric loading of body weight and external loading is reduced as the perpendicular components of force become dominant. Both collective and distributed loading patterns provide thrustlines of approximately the same depth of curvature. However, the addition of muscle forces which flatten the overall curvature and diminish the effects of external loading may restore the smoother curved representation of the thrustline obtained with the distributed loading pattern. Also, when muscle forces are added, the small difference in moments due to body weight may be the difference between a total positive or negative moment and result in a different direction for the end support reaction forces.
5.1.2 THE EFFECTS OF DISTRIBUTED BODY WEIGHT, MUSCLE AND LIGAMENT FORCES ON PREDICTED THRUSTLINES

Comparison of the predicted thrustlines for collective and distributed loading patterns was continued for the application of muscle forces. For the collective loading pattern a single force representing muscle and ligament forces was applied at the upper end of the spine. The value of this force was varied until the thrustline which provided the closest fit with the path of the vertebral bodies was obtained. The best fit thrustline was defined as that which represented the minimum sum of the distance between the thrustline and spine at each vertebral level. The predicted thrustlines were then compared with the closest fit thrustline determined using the distributed loading patterns in sections 4.6.2 and 4.6.3.

5.1.2.1 Direction of the Thrustline Path

With a collective loading pattern the projection of the thrustline due to body weight follows a posterior convex curve. The applied muscle force at the upper end of the spine flattens the overall thrustline but cannot alter the curve locally. Figure 5-5 shows the predicted thrustline for the erect posture which provides the closest fit thrustline for these loading conditions. This runs close to the centre of the vertebral bodies in the thoracic region but does not coincide with the alternating curves of the lumbar and cervical regions.
In contrast, with the distributed loading pattern the application of individual muscle forces in certain regions of the spine, with perpendicular and parallel components allows the path of the thrustline to be controlled locally. This is demonstrated for the erect posture in Figure 5-5 by a combination of lumbar multifidus (activated to 20%), intertransverse and intra abdominal pressure (4KPa) forces, which produce localised flattening of the cervical and lumbar curves.

The application of muscle forces at certain levels of the spine therefore allows the local control of the thrustline which cannot be provided by a collective force. This is necessary to accommodate for the variations in curvature which can be achieved with flexion and extension at each individual vertebral level.

Figure 5-5: Illustrating the closest fit thrustline determined for the collective and distributed loading patterns for body weight and muscle and ligament forces in the erect posture.
5.1.2.2 Indication of Stability

In the flexed posture, the configuration of the spine tends towards that of a single posterior convex curve. However, the thrustline predicted for the collective body weight and muscle forces coincides with the path of the vertebral bodies at its points of force application, but passes outside the vertebral bodies at the thoracic levels between these points (Figure 5-6). While the actual shape of the curve is due to the collective loading for body weight, the collective muscle force can only increase or decrease the depth of this curve. In contrast, muscle forces applied locally allow more precise control of the shape of the curve.

Figure 5-6 shows the smoother thrustline curve predicted for forces applied due to the cervical muscle semispinalis capitis combined with either lumbar muscle iliocostalis lumborum and multifidus. This satisfied conditions of stability at all spinal levels.
Figure 5-6: Comparison of the closest fit thrustlines determined using collective and distributed loading patterns for body weight and muscle and ligament forces in the flex2 posture.

5.1.2.3 Requirements for Muscle Force

Representation of the forces due to each individual muscle group reveals areas of instability where the muscle and ligament forces may be insufficient to support an external load.
The predicted thrustline for the collective loading in the flex2 posture satisfied stability at every level (Figure 5-7). In contrast, for the distributed loading pattern complete activation of the greatest spine extensor muscle forces (longissimus thoracis, thoracic iliocostalis and semispinalis capitis) could not satisfy stability at the upper regions of the spine. Similar results were also found for the flex3 posture. Further study indicated the necessity for shoulder muscle activation. Using the distributed muscle force pattern therefore indicates essential roles for individual muscle groups.

Figure 5-7: Illustrating the closest fit thrustline determined for the collective and distributed loading patterns for body weight, external load and muscle forces in the flex2 posture.
5.1.2.4 Amount of Muscle Force Required.

A collective muscle force applied at the upper end of the spine assumes the muscle forces act in a single direction over the entire length of the spine.

However investigation of the individual muscle groups showed some muscles such as cervical and thoracic multifidus, iliocostalis cervicis, scalenus posterior and psoas to increase the curvature of the thrustline due to body weight in the erect posture. Muscle forces do therefore not necessarily result in a flattening effect on the thrustline. The contribution of the muscle forces therefore requires consideration of the direction of each individual force. Also the muscle forces generated by each individual group act over different levels of the spine. This consequently affects the amount of force required.

Table 5-3 shows the amount of force required for each posture, under the application of body weight alone or with an external load of 900N when a collective muscle force is applied. A comparison is also made with an equivalent lever model in which the muscle force acts parallel to the L5/S1 joint with a moment arm of 5cm.
Larger values of muscle force are required for the erect posture with the arch model than with the lever model. This is due to lordotic curves of the cervical and lumbar vertebrae in this posture, whereby the function defining the minimum distance between vertebrae and thrustline requires a greater force. Also greater force is required for the arch model in the flex1 posture in comparison with the flex2 posture for this reason. The greater muscle force required in the flex3 conditions reflects the greater perpendicular force components of body weight. However the force required in the flexed postures is less than that for the lever model.

Table 5-3: Collective muscle and ligament force required to obtain the closest fit thrustline with the vertebrae under body weight and externally applied load.

<table>
<thead>
<tr>
<th>Body Weight</th>
<th>Collective Muscle Force Required (N) - Arch Model</th>
<th>Collective Muscle Force Required (N) - Lever Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erect</td>
<td>600</td>
<td>36.5</td>
</tr>
<tr>
<td>Flex1</td>
<td>280</td>
<td>462</td>
</tr>
<tr>
<td>Flex2</td>
<td>165</td>
<td>766.8</td>
</tr>
<tr>
<td>Flex3</td>
<td>328</td>
<td>1175.8</td>
</tr>
<tr>
<td>Body Weight and</td>
<td></td>
<td></td>
</tr>
<tr>
<td>External Load</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Erect</td>
<td>950</td>
<td>285.40</td>
</tr>
<tr>
<td>Flex1</td>
<td>1165</td>
<td>1957.60</td>
</tr>
<tr>
<td>Flex2</td>
<td>755</td>
<td>3942.00</td>
</tr>
<tr>
<td>Flex3</td>
<td>1330</td>
<td>5881.80</td>
</tr>
</tbody>
</table>
The distributed loading pattern for muscle forces involves some force pairs in opposing directions which make it difficult for a direct comparison with the collective muscle force. In the erect posture the force set (20% lumbar multifidus, intertransverse, intra abdominal pressure) was predicted to provide the closest fit between thrustline and vertebrae. For the multifidus (20%), 24.2N is applied as opposing force pairs, and 121N as single forces, and for the intertransverse 450N is applied in opposing force pairs. Intra abdominal pressure of 10N applied perpendicular to each lumbar vertebra and 4N to each intervertebral disc is also needed. This demonstrates a more complex loading pattern with greater diversity of force directions.

Table 5-4 summarises the total force generated for each muscle combination identified to provide a close fit thrustline due to body weight and an external load for each posture. In comparison with the collective muscle force, the individual muscles groups which satisfy stability involve several force pairs. Even the lumbar muscles longissimus thoracis and multifidus have some fascicles attaching at both ends to the vertebrae which therefore act in opposing directions. These muscle groups with attachments between the ends A and B can provide a flattening of the thrustline curve without affecting the end support components H1 and H2. Thus whereas for the collective muscle force the opposing sacral reaction (H1) provides the compression at the opposite end of the spine, using individual muscle groups applies opposing force pairs at smaller intervals between H1 and H2. Due to the closer application of these forces, the amount of force required is greater. This was illustrated for the posterior longitudinal ligament for which, despite the generation of large forces, the overall effect on the thrustline is small.
Table 5-4: Force Required by Combinations of Individual Muscle and Ligament Groups to Obtain the Closest Fit Thrustline with the Vertebrae Under Body Weight and Externally Applied Load

<table>
<thead>
<tr>
<th>Force Combination</th>
<th>Force Required (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Erect + Body Weight</strong></td>
<td>lumbar multifidus in intra abdominal pressure (4KPa) intertransverse 121N force pair, 725N single 750N force pair</td>
</tr>
<tr>
<td><strong>Flex1 + Body Weight</strong></td>
<td>Intra abdominal pressure (6KPa) 55% lumbar iliocostalis 25% semispinalis capitis 75N vertebrae, 30N discs 348.15 single direction 135N force pair</td>
</tr>
<tr>
<td><strong>Flex2 + Body Weight</strong></td>
<td>semispinalis capitis iliocostalis lumbarum or multifidus 540 force pair 633 single direction 121N force pairs, 725N single direction</td>
</tr>
<tr>
<td><strong>Flex3 + Body Weight</strong></td>
<td>25% longissimus thoracis semispinalis capitis 231 force pairs, 1577 single direction 540 force pairs</td>
</tr>
<tr>
<td><strong>Erect+ External Load</strong></td>
<td>50% longissimus capitis iliocostalis cervicis intertransverse IAP (12KPa) 40N force pair 450N force pair, 210N IAP</td>
</tr>
<tr>
<td><strong>Flex1+ External Load</strong></td>
<td>longissimus thoracis (curved) lumbar multifidus semispinalis capitis IAP (14KPa) 231 force pairs, 1577 single direction 121N force pair, 725N single 540 force pairs, 245N</td>
</tr>
<tr>
<td><strong>Flex2 and Flex3 + External Load</strong></td>
<td>longissimus thoracis (curved) thoracic iliocostalis (curved) semispinalis capitis 231 force pairs, 1577 single direction 547 single direction 540 force pairs</td>
</tr>
</tbody>
</table>
5.1.2.5 Summary

Representation of the individual muscle groups acting along the spine is necessary for four main reasons:

1) The variation in spinal curvature at each level requires local control of the thrustline which cannot be provided by a collective muscle force. This is particularly important for ensuring stability of the spine in the erect posture.

2) Using individual muscle group forces reveals essential roles for particular muscle groups such as the shoulder muscles in the flex2 and flex3 postures when supporting a 900N external load which are not found using a collective muscle force.

3) Investigation of the individual muscle groups allows the effects of both perpendicular and parallel components of force to be considered. Depending upon the force direction, this may flatten or increase the thrustline curve, whereas a collective muscle force applied parallel to the reference line predicts a flattening effect.

4) A collective muscle force applied at the upper end of the spine takes effect over the whole spinal column. However the contribution of the ligaments and muscles acting over smaller intervals and require greater force to be applied.
5.1.3 THE EFFECTS OF DISTRIBUTED BODY WEIGHT FORCES ON PREDICTED JOINT REACTIONS

The path of the thrustline indicates the direction of the resultant force at each level of the spine. Close alignment between the thrustline and the vertebral longitudinal axis indicates the force is mainly compression, and the corresponding shear component is small. Due to the difference in thrustline paths predicted with the collective and distributed loading, a comparison between the joint reactions predicted for collective and distributed body weight loading patterns was made.

5.1.3.1 Observations Common to both Loading Patterns

5.1.3.1.1 Intervertebral Compression

The predicted compression values for each of the four postures are shown in Figure 5-8 for the distributed loading case and Figure 5-9 for the collective loading case.
Figure 5-8: Compression values at each intervertebral level for a distributed body weight loading pattern for the various postures. (Joint level 1 = L5/S1, joint level 24 = C1/C2.)

Figure 5-9: Compression values at each intervertebral level for a collective body weight loading pattern for the various postures.
Both figures show an increase in compression descending the spine reflecting the cumulative effect of body weight forces. Also the values are all positive, indicating the reaction force is directed upwards to satisfy equilibrium with the downward component of body weight.

Both loading patterns also show the compression is greatest in the erect posture, and least in the flexed posture. In the erect posture depth of the thrustline curve is small, and the path of the thrustline relative runs closer to the joint axes resulting in a greater compression component. In contrast in flexion the depth of the thrustline curve is increased considerably, thereby increasing the shear components of force and decreasing the corresponding compression (Figure 5-10).

An exception is at the upper cervical levels of the spine, where the compression for the erect posture at these levels is less than that for the flexed postures. This is due to the values of the end support reactions for these postures. At lower levels the addition of body weight increases the parallel component and results in a greater compression component in the erect posture, and a smaller component in the flexed posture as above.
Figure 5-10: The path of the resultant forces due to body weight relative to the spine in the erect and flex3 postures, to illustrate compression and shear components.
5.1.3.1.2 Intervertebral Shear

The predicted values of shear for the distributed and collective loading patterns are shown in Figure 5-11 and Figure 5-12 respectively. The figures show an increase in shear components with flexion, which corresponds to the decrease in compression explained above. However, whereas the compression values are all positive, the value of the shear component are both positive and negative. Figure 5-10 shows the path of the resultant forces with positive shear components at the upper vertebral levels. At the apex of the curve the thrustline then changes direction resulting in negative shear components at the lower vertebral levels. The joint reaction forces are therefore negative at the upper vertebral levels and positive at the lower vertebral levels, with very low shear values at the transition in the upper thoracic regions (Figure 5-11).

At the lowest lumbar levels, the greatest shear component is for the flex1 posture, and least for the flex2 posture, contradicting the increase in shear observed with flexion. This is due to the direction of the end support reaction components at this end, directing the thrustline closer to the vertebral axes for the erect and flex3 postures.
Figure 5-11: Shear values at each intervertebral level for a distributed body weight loading pattern for the various postures.
Figure 5-12: Shear values at each intervertebral level for a collective body weight loading pattern for the various postures.

5.1.3.2 Distinction between Collective and Distributed Loading Patterns

Figure 5-9 and Figure 5-12 show sharp changes in compression and shear values respectively at the vertebrae corresponding to the level of applied loading. In contrast Figure 5-8 and Figure 5-11 show a gradual change in compression and shear reflecting the application of a smaller load at each level.

The difference is particularly noticeable in the lumbar regions where a gradual increase in positive shear values occurs descending the lumbar spine for the distributed loading case due to the gradual application of force. In contrast for the collective force, a single load of large magnitude representing the weight of the trunk is applied at T12 (joint level 6) which results in larger compression and shear values at all the lumbar levels.
Changes in vertebral dimensions at each level of the spine reflect the load bearing capacity. Descending the spine a gradual increase in vertebral body cross sectional area indicates a greater load bearing capacity at the lumbar levels. The distributed loading pattern is considered to provide a closer match between vertebral compressive and shear loading, and these changes in vertebral dimensions at each level.
5.1.4 THE EFFECTS OF DISTRIBUTED BODY WEIGHT AND MUSCLE AND LIGAMENT FORCES ON PREDICTED JOINT REACTIONS

The predicted joint reactions for the distributed and collective loading cases for body weight and muscle forces which provided the closest fit between the path of the vertebral bodies and the thrustline were compared.

5.1.4.1 Compression

5.1.4.1.1 Erect Posture

In the erect posture, the combination of intra abdominal pressure, lumbar multifidus activated to 20% and intertransverse for the distributed loading case was compared with a collective muscle force of 280N. The compressive values due to a collective muscle force pattern were much greater than those for a distributed force pattern. The predicted compression at the L5/S1 joint for the collective loading pattern was 897.25N, and for the distributed loading pattern 509.24N. The discrepancy between the predicted values for the loading patterns is due to the algorithm used to determine the best fit thrustline for the collective loading case. For the collective loading the overall thrustline was flattened whereas for the distributed case the thrustline was flattened in only the cervical and lumbar regions. The amount of muscle force applied and the consequent joint reactions were therefore less for the distributed case.
5.1.4.1.2 Flex2 Posture and Body Weight

Comparison of the joint reactions was considered more important in the flexed postures for which good fit thrustlines for both loading cases had been obtained. Figure 5-13 shows the predicted compression for the collective loading case, and for combinations of semispinalis with iliocostalis lumborum or multifidus, and submaximal forces for semispinalis and longissimus thoracis in the flex2 posture.

The latter combination of submaximal forces results in lower predicted compression due to the application of forces of lower magnitude than for the fully activated combinations. However greater compression was predicted in the thoracic regions in which these forces were applied. In contrast combinations of semispinalis capitis and iliocostalis or multifidus resulted in greater compression in the cervical and lumbar regions in which they were applied, and lower compression at the thoracic levels.

Due to the lower values of compression predicted for the longissimus thoracis and semispinalis muscle combination, this would be a preferable recruitment strategy. Also these muscles are only partially activated, and would therefore not fatigue as quickly as combinations of semispinalis and lumbar iliocostalis or multifidus fully activated.

The peak compression value predicted for the collective loading pattern of 456.4N at the L1/L2 level is therefore to be compared with a value of 595.3N for longissimus thoracis and semispinalis capitis at the L4/L5 level. The value of 313.64N at the C1/C2 level must also be noted for this combination as values in the cervical region are also higher than for the collective loading case.
5.1.4.1.3 Flex3 Posture and Body Weight

The predicted compression values for the flex3 posture are shown in Figure 5-14. For the distributed loading a combination of 25% longissimus thoracis and semispinalis capitis resulted in greater compression at the upper and lower levels of the spine. At the L5/S1 level this was 578.16N, but more significantly at the C2/C3 level compression was 547.31N. In contrast the collective muscle force applied at the upper end of the spine resulted in greater values of compression at the thoracic levels, but lower values at the cervical levels. This type of loading pattern could therefore not predict the variation in loading at each level due to the application of individual muscle groups.
5.1.4.1.4 Flex1 Posture and External Load

In the flex1 posture the addition of an external load required greater muscle force to ensure stability. For the collective loading case due to the application of the total muscle force at the upper end of the spine, forces at the cervical and thoracic levels were high (Figure 5-15). These were greater than the combination of extensor muscle forces used in the distributed loading pattern which applied the greatest force at the lumbar levels.
Figure 5-15: Comparison of predicted compression for collective and distributed muscle force loading patterns in the flex3 posture.

5.1.4.1.5 Flex3 Posture and External Load

In the flex3 posture, to support an external load the distributed loading pattern required the greatest amount of muscle force and therefore represents the case with greatest loading on the spine for the four postures investigated.
The predicted peak compression value for the collective loading was 1767.12N at the L1/L2 level. Using the distributed loading, the combination of extensor muscles which provided the closest fit thrustline (longissimus thoracis, thoracic iliocostalis and semispinalis capitis) resulted in a greater peak compression value of 2711.05N at the L5/S1 level. Also, considerably lower values were predicted at the cervical and thoracic levels than for the collective muscle force when applied at the upper end of the spine. With the addition of shoulder muscles the compression at thoracic and lumbar levels was increased (Figure 5-16).

![Graph](image-url)

Figure 5-16: Comparison of predicted compression for collective and distributed muscle force loading patterns to support an external load in the flex3 posture.
5.1.4.2 Shear

In the erect and flex1 postures, a closer fit thrustline was predicted for the distributed loading than for the collective loading. For the distributed loading pattern the direction of the resultant forces therefore corresponded mainly to the direction of the intervertebral joint axes against which compression was calculated and the corresponding components of shear were smaller. This is shown for the erect posture in Figure 5-17. The shear values are particularly high at the lower lumbar levels where the predicted thrustline for the collective loading pattern is projected in the opposite direction to the path of the vertebral bodies (Figure 5-5).

![Figure 5-17: Comparison of predicted shear for collective and distributed muscle force loading patterns for the erect posture.](image)
For the flex3 posture, close fit thrustlines were found for both loading patterns. However the distributed loading pattern resulted in a smoother thrustline curve which corresponded more closely to change in vertebral axis direction. In contrast the thrustline for the collective loading pattern showed abrupt changes in direction. Shear values for this case were therefore greater due to difference in direction between the thrustline and vertebral axes at the thoracic levels (Figure 5-18).

Figure 5-18: Comparison of predicted shear for collective and distributed muscle force loading patterns for the flex3 posture.
With the application of an external load, the magnitude of shear components for both loading patterns are greater than those due to body weight alone due to a resultant force of greater magnitude. The flex3 posture was considered to represent the case requiring greatest muscle force and therefore the predicted joint reactions represent the maximum of all postures. The predicted values are shown in Figure 5-19. The greatest shear value was 769N at the T5/T6 level for the combination of extensor muscles (longissimus thoracis, thoracic iliocostalis and semispinalis capitis). With the addition of shoulder muscles the thrustline was moved closer to the path of the vertebral bodies, and the shear values were therefore reduced.

Figure 5-19: Comparison of predicted shear for collective and distributed muscle force loading patterns for the flex3 posture with the addition of an external load.
5.1.4.3 Discussion

5.1.4.3.1 Compression

For the collective loading, the single force representing the action of muscles and ligaments applied at the upper end of the spine acts at every vertebral level. The components of compression and shear for the predicted joint reactions at every level of the spine are therefore increased by the same amount. The overall pattern for the predicted joint reactions for body weight is therefore followed. Realistically because the collective force represents the total muscle and ligament force required to stabilise the spine in a given posture, only the predicted reaction at the lower lumbar intervertebral joints should be considered.

In contrast, for the distributed loading pattern, application of forces for individual muscle groups results in localised effects on the predicted joint reactions. This type of loading can therefore be used to consider the loading at each individual level of the spine. Both the compression at the L5/S1 joint, and the peak compression values at the level at which they occurred were therefore considered. These are compared with the compression at the L5/S1 level for the collective loading in Table 5-5. A comparison is also made between these predictions and those for a traditional lever model in which a collective muscle force acting 5cm posterior to the L5/S1 joint in a parallel direction was considered.
Table 5-5: Predicted peak compression values (N) for the arch model with collective and distributed loading, and the lever model with collective loading.

(sscapitis=semispinalis capitis, longthor=longissimus thoracis)

<table>
<thead>
<tr>
<th>Posture and Muscle Combination (distributed loading)</th>
<th>Arch Model, Distributed Loading (Various levels)</th>
<th>Arch Model, Collective Loading (L5/S1 level)</th>
<th>Lever Model, Collective Loading (L5/S1 level)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Body Weight</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Erect: IAP (4Kpa) + multifidus</td>
<td>509.24 (L5/S1)</td>
<td>897.25</td>
<td>431.61</td>
</tr>
<tr>
<td>Flex1: IAP (6Kpa) + 55% iliocostalis + 25% sscapitis</td>
<td>690.66 (L4/L5)</td>
<td>589.01</td>
<td>836.32</td>
</tr>
<tr>
<td>Flex2: 20% longthor + 50% sscapitis</td>
<td>595.3 (L4/L5)</td>
<td>441.91</td>
<td>1068.03</td>
</tr>
<tr>
<td>Flex3: 25% longthor + sscapitis</td>
<td>578.16 (L5/S1)</td>
<td>478.76</td>
<td>1323.76</td>
</tr>
<tr>
<td>Body Weight+ External Load</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Erect: 50% longissimus capitis + IAP (12Kpa) + iliocostalis cervicis</td>
<td>1282.02 (L3/L4)</td>
<td>2029.75</td>
<td>1580.03</td>
</tr>
<tr>
<td>Flex1: IAP (14KPa) + longthor + sscapitis + multifidus</td>
<td>2857.19 (L4/L5)</td>
<td>2211.02</td>
<td>3224.11</td>
</tr>
<tr>
<td>Flex2: longthor + sscapitis + thoracic iliocostalis</td>
<td>3117.49 (L4/L5)</td>
<td>1670.27</td>
<td>4929.03</td>
</tr>
<tr>
<td>Flex2: longthor + sscapitis + thoracic iliocostalis + shoulder muscles</td>
<td>3018.79 (L5/S1)</td>
<td>1670.27</td>
<td>4929.03</td>
</tr>
<tr>
<td>Flex3: longthor + sscapitis + thoracic iliocostalis</td>
<td>2711.05 (L5/S1)</td>
<td>1761.07</td>
<td>6366.62</td>
</tr>
<tr>
<td>Flex3: longthor + sscapitis + thoracic iliocostalis + shoulder muscles</td>
<td>3634.16 (T7/T8)</td>
<td>1761.07</td>
<td>6366.62</td>
</tr>
</tbody>
</table>
The predicted compression values for the arch model with both collective and distributed loading are lower than the predicted values for the lever model, for all cases except the erect posture.

Joint reaction predictions for the arch model using the distributed loading pattern lie between the values reported for the lever model and the arch model with collective loading.

In the flex1 and flex3 postures greater compression in the cervical and lumbar levels was predicted using the distributed loading pattern. This was due to the application of individual muscle groups in this area. The collective loading pattern cannot therefore predict the variation in joint reaction forces which occurs with the action of individual muscle groups.

Also the application of a single force at the upper end of the spine for the collective loading results in an increase in compression at every level. Consequently in the flex3 posture the predicted compression for the collective loading was higher at thoracic levels than for the distributed loading.

For the flex3 posture the predicted compression at the L5/S1 joint was 3062.48N when supporting the external load. This is to be compared with 1761.07N predicted for the collective loading. When considering the action of additional shoulder muscles, the additional force is realistically distributed over the shoulder girdle rather than directly applied on the vertebrae in this case. The compression may be lower. However even without the shoulder muscles the predicted compression of 2711.05N is still 54% greater than that predicted for the collective loading.
Comparison with the joint reaction predictions for a lever model with greater anatomic detail is difficult due to the diversity of loading conditions reported. McGill et al. (1986) reported the predicted compression at the L4/L5 joint for subjects lifting a 91Kg (891.80N) load ranged from 6264N-8921N. The equipment used constrained the loads to be lifted in the vertical plane, similar to lifting strategy modelled in this work. However these values were reported for dynamic lifting, which requires greater force to overcome the inertia of the trunk. Despite this the authors reported a reduction in compression by 8.9-9.1% when compared with the collective loading. In contrast, this model predicts a 54% increase in compression when individual muscle forces are considered for an arch model.

In the study of McGill et al. (1986) the main trunk extensor and flexor muscles were included, with the erector spinae and psoas muscles each represented by a number of fascicles. A later study by Cholewicki and McGill (1995) developed the anatomic detail further to include a total of 50 muscle fascicles. Subjects performed lifts to near maximum effort and the trunk extension torque recorded by a dynamometer was 171Nm. A combined EMG-optimisation based model predicted the mean compression to be 3984N for the dynamic lift performed. This value is considerably closer to that reported in the current model.
The use of EMG data as input to these models accounted for abdominal muscle coactivity. This study considered activity for rectus abdominis (5% activation level), and the oblique muscles (10% activation), along with the necessary increase in lumbar muscle force to counteract the increase in flexion moment for the flex2 posture. The consequent compression at the L5/S1 joint was increased from 3018.79N to 4433.73N; a 47% increase. This value may be lower when shoulder muscle force is considered distributed over the shoulder girdle. Values between the detailed lever model and this detailed arch model may therefore be in closer agreement. However the difference in model testing conditions makes it difficult to draw conclusions. Certainly representation of the spine as an arch does not result in the significant reduction in joint compression reported by Aspden (1987) when individual muscle groups are included.

In all postures, the predicted compression values using the collective and distributed loading patterns for the arch model are significantly lower than the known failure values of the lumbar vertebrae of 5-8KN reported by Farfan, 1973. In contrast for the lever model in the flex3 posture when supporting a 900N load, the predicted compression of 6366.2N corresponds to the failure range of the lumbar vertebrae.

However the distributed loading pattern indicates loading at other levels must also be considered. For the flex1 and flex2 postures, the peak compression occurred at the L4/L5 levels. For the flex3 posture high compression values were also predicted for the cervical regions. Also for the flex3 postures when supporting an external load, the inclusion of the shoulder muscles increased the compression at the thoracic levels.
In Table 5-6 the values for vertebral compressive strength are obtained from Goel et al. (1990). The lower limit for the lumbar vertebrae is less than that reported by Farfan (1973), indicating considerable biological range among specimens. When supporting body weight only, the reported compression values are well below the lower limit for vertebral end plate failure. When considering the addition of an external load, the compression values are above the lower limit for the reported failure values for both loading cases, but are well below the upper limit. It is therefore reasonable to conclude that the increase in vertebral compression predicted by this model is unlikely to result in vertebral compressive failure.
Table 5-6: Comparison of the range of compression (N) predicted for each spinal region with reported failure values for the vertebral end plates (Goel et al., 1990).

<table>
<thead>
<tr>
<th>Body Weight and External Load</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar (without shoulder muscles)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Body Weight</strong></td>
<td>Erect</td>
<td>Flex1</td>
<td>Flex2</td>
</tr>
<tr>
<td>Cervical</td>
<td>592.47-591.19</td>
<td>282.86-284.71</td>
<td>313.64-199.74</td>
</tr>
<tr>
<td>Distributed</td>
<td>60.40-49.46</td>
<td>164.99-118.82</td>
<td>197.55-197.96</td>
</tr>
<tr>
<td><strong>Body Weight and External Load</strong></td>
<td>Collective</td>
<td>582.57-940.47</td>
<td>283.93-625.80</td>
</tr>
<tr>
<td>Distributed</td>
<td>74.79-273.73</td>
<td>126.53-298.82</td>
<td>198.03-456.32</td>
</tr>
<tr>
<td>Lumbar</td>
<td>949.50-897.25</td>
<td>629.53-589.01</td>
<td>501.95-583.14</td>
</tr>
<tr>
<td>Distributed</td>
<td>349.94-509.24</td>
<td>421.22-680.76</td>
<td>456.44-441.91</td>
</tr>
</tbody>
</table>

Body Weight Erect Flex1 Flex2 Flex3 Reported failure values
Cervical 592.47-591.19 282.86-284.71 313.64-199.74 343.61-358.30 667-4450
Distributed 60.40-49.46 164.99-118.82 197.55-197.96 346.08-319.59

<table>
<thead>
<tr>
<th>Body Weight and External Load</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Body Weight and External Load</strong></td>
<td>Collective</td>
<td>940.11-933.40</td>
<td>1184.78-1183.59</td>
</tr>
<tr>
<td>Lumbar</td>
<td>918.80-2148.28</td>
<td>1179.38-2324.74</td>
<td>877.77-1691.96</td>
</tr>
<tr>
<td>Distributed</td>
<td>156.73-153.13</td>
<td>402.71-1954.97</td>
<td>449.10-2709.33</td>
</tr>
<tr>
<td>Lumbar (without shoulder muscles)</td>
<td>2167.83-2029.75</td>
<td>2340.07-2211.02</td>
<td>1679.11-1670.27</td>
</tr>
<tr>
<td>Distributed</td>
<td>1201.90-1234.58</td>
<td>2254.99-2703.30</td>
<td>3117.49-2871.04</td>
</tr>
</tbody>
</table>

416
5.1.4.3.2 Shear

Lower values of shear are observed at levels of the spine which show a close fit between the path of the vertebrae and the resultant forces. Table 5-7 reports the peak shear values for the collective and distributed loading patterns which provided the best fit thrustlines. Values are highlighted where the predicted shear for the distributed case is greater than that for the collective loading.

Table 5-7: Predicted peak shear (N) values (antero-posterior and postero-anterior) and the corresponding intervertebral level for the arch model with collective and distributed loading.

<table>
<thead>
<tr>
<th>Body Weight</th>
<th>Peak Antero-posterior Shear (positive)</th>
<th>Peak Postero-anterior Shear (negative)</th>
<th>C1/C2 shear</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Collective</td>
<td>Distributed</td>
<td>Collective</td>
</tr>
<tr>
<td>Erect</td>
<td>392.82</td>
<td>134.54</td>
<td>-275.06</td>
</tr>
<tr>
<td></td>
<td>(L5/S1)</td>
<td>(L5/S1)</td>
<td>(T12/L1)</td>
</tr>
<tr>
<td>Flex1</td>
<td>242.02</td>
<td>184.26</td>
<td>-133.70</td>
</tr>
<tr>
<td></td>
<td>(L5/S1)</td>
<td>(L5/S1)</td>
<td>(T11/T12)</td>
</tr>
<tr>
<td>Flex2</td>
<td>115.76</td>
<td>157.58</td>
<td>-94.88</td>
</tr>
<tr>
<td></td>
<td>(L5/S1)</td>
<td>(L5/S1)</td>
<td>(T11/T12)</td>
</tr>
<tr>
<td>Flex3</td>
<td>147.34</td>
<td>88.21</td>
<td>-111.87</td>
</tr>
<tr>
<td></td>
<td>(T2/T3)</td>
<td>(L5/S1)</td>
<td>(C1/C2)</td>
</tr>
<tr>
<td>Body Weight + External Load</td>
<td>Collective</td>
<td>Distributed</td>
<td>Collective</td>
</tr>
<tr>
<td>Erect</td>
<td>923.18</td>
<td>380.57</td>
<td>-597.5</td>
</tr>
<tr>
<td></td>
<td>(L5/S1)</td>
<td>(L5/S1)</td>
<td>(T12/L1)</td>
</tr>
<tr>
<td>Flex1</td>
<td>863.73</td>
<td>868.27</td>
<td>-480.86</td>
</tr>
<tr>
<td></td>
<td>(L5/S1)</td>
<td>(L5/S1)</td>
<td>(T11/T12)</td>
</tr>
<tr>
<td>Flex2 (without shoulders)</td>
<td>279.85</td>
<td>783.02</td>
<td>-249.99</td>
</tr>
<tr>
<td></td>
<td>(L5/S1)</td>
<td>(T4/T5)</td>
<td>(L3/L4)</td>
</tr>
<tr>
<td>Flex3 (without shoulders)</td>
<td>710.62</td>
<td>496.75</td>
<td>-483.65</td>
</tr>
<tr>
<td></td>
<td>(T6/T7)</td>
<td>(T4/T5)</td>
<td>(C1/C2)</td>
</tr>
</tbody>
</table>
Due to the application of individual muscle forces, the vertebral level at which the peak shear values occur varies. The collective loading pattern which applies a single muscle force at the upper end of the spine does not account for individual variation in muscle force which occurs when stabilising each intervertebral level.

The shear values in the flex2 and flex3 postures with external load show two cases for which the shear values are greater for the distributed loading, due to the closer fit thrustline obtained for the collective loading. (Peak antero-posterior shear is higher in the thoracic region in the flex2 posture and peak postero-anterior shear is higher in the lumbar region in the flex3 posture.) However for the majority of cases the predicted shear values are much less with the distributed loading pattern, due to the direction of thrustline corresponding more closely with the inter vertebral joint longitudinal axes.

Values at the C1/C2 level represent the combined force due to the end support reaction R2, and perpendicular components of force due to head weight and neck muscles. These are considerably lower using the distributed loading pattern for all but one cases. The distributed loading pattern thereby reduces the force applied at the upper end of the spine.
5.1.5 REPRESENTATION OF THE WHOLE SPINE

Consideration of the stability of the whole spine has shown a number of cases in which the thrustline coincides with the path of the vertebral bodies in the lower regions of the spine, but not in the upper regions. In particular, in the flex2 and flex3 postures, extensor muscle activity was sufficient to ensure stability of the thoraco-lumbar spine but not in the additional cervical regions. Stability of the lumbar or thoraco-lumbar spine does therefore not represent stability of the whole spine.

Alteration in the applied muscle force pattern to ensure whole spine stability alters the predicted joint reaction forces and the total amount of muscle force required. Cervical muscle forces which may be added to ensure cervical spine stability also increase the overall flattening of the thrustline and may result in instability in the thoracic and lumbar regions. Force conditions which satisfy partial spine stability is therefore not a realistic representation of spinal loading in vivo. Consideration of the whole spine is therefore necessary to obtain a more realistic estimate of the loading in each region associated with a given posture.

5.1.6 METHOD OF INVESTIGATION

Combinations of muscle groups which satisfied stability for each posture were determined based upon the understanding of their individual behaviour.

An alternative approach would be to use an optimisation procedure, whereby combinations of the unknown muscle forces which minimise the distance between the thrustline and centre line of the arch at each level are determined.
By considering various levels of activation this method could automatically investigate a greater number of possible combinations. However, this could also yield a vast amount of combinations which do not represent the muscle recruitment pattern for an individual.

At these early stages of model development and investigations it was considered more important to obtain an understanding for the contribution of each muscle to spinal stability and to identify combinations which realistically reflected the muscle roles. Consequently muscle groups important for stability of the spine have been identified and can be included in more detailed future investigations.

5.1.7 MODEL PERFORMANCE

5.1.7.1 Program Run Time

The collective loading used by Aspden (1987) involved the application of only three external loads. These can be arranged manually so that they are in sequential order. The computer code therefore only needs to calculate the moments generated about the end supports, to predict the end support reactions and to calculate the resultant force components for four links of the thrustline.
Application of muscle and body weight forces using a distributed loading pattern involves a considerable increase in the number of forces to be considered. Twenty four forces are involved with the distributed body weight pattern. These are entered in the model in sequential order so that sorting of the force markers is not required. However, the addition of a muscle group requires the muscle force markers be slotted into the sequence of body weight forces. To provide a choice for the user for the muscle group to be tested, it is required that any muscle group can be added. The pre-arranging of markers into sequential order manually would therefore be time consuming. A sorting procedure is therefore performed by the computer.

The sorting procedure involves systematically comparing the location of each marker with the location of every other marker. This requires a nested loop and involves \(2^n\) iterations (\(n\) is the number of forces). The time to perform a run is therefore increased considerably and the requirement for a more precise anatomic model is compromised by the time to perform an analysis.

5.1.7.2 Alteration of Parameter Values

Positioning of the vertebrae relative to each other within the program results in a series of parts (vertebrae) all dependent upon the position and orientation of the sacrum. Further, the positions and directions of markers and the values for variables which represent geometry and force properties are dependent upon the vertebrae to which they belong. From this information the predicted thrustline is calculated.
Alteration of an applied muscle force value, or a change in vertebra position therefore requires recalculation of the thrustline and joint reactions. Due to the complex dependence of the variables upon markers, markers upon parts and parts sequentially upon each other, re-calculation of these variables in turn requires all other variables are updated. This takes an incredibly long time! It is therefore preferable to re-run the simulation using the new parameter values.
5.2 MODEL VALIDATION

5.2.1 JOINT REACTION PREDICTIONS AND INTRA DISCAL PRESSURE MEASUREMENTS

5.2.1.1 Erect Posture

In the erect posture, the predicted compression at the L5/S1 intervertebral joint when supporting body weight using the distributed loading patterns was 431.61N. A reasonable agreement is found with the values reported by Dolan and Adams (1995) in quiet standing of 500N.

Also Schultz et al. (1982) predicted an intra discal pressure of 270 KPa during relaxed standing. Assuming the cross sectional area of the disc corresponds to that of the L5 vertebral body, the dimensions applied in the current model yield an area of 1669.32mm². Assuming the applied pressure is carried by the entire disc provides an indication of the compression at this level. Using the relation force = pressure * area, the predicted compression at the disc was 450.72N. This is likely to be slightly greater as approximately only 70% of the load is carried by the nucleus (Aspden, 1989). However, it provides close comparison with the predicted joint reactions in the current model.
5.2.1.2 Flexed Postures

In vivo measurements for intra discal pressure were performed by Schultz et al. (1982). For subjects with a trunk flexion angle of 30° and with the arms held out in front, the intra discal pressure was 1040Kpa. This corresponds to a compressive force of at least 1736.09N following the above calculations.

The flex1 and flex2 postures investigated in the current study represent trunk flexion angles of 18.76° and 40.36° respectively. For these postures the predicted compression at the L5/S1 joint was 578.16N and 1323.76N. These postures which represent trunk flexion angles above and below 30°, predict compressive lower than the value reported by Schultz et al. (1982). However, while the combinations of muscle forces were found to satisfy stability, they may not be indicative of the actual muscle recruitment patterns. Close agreement between the model predictions and in vivo measurements are therefore not necessary. More importantly the model predictions do not exceed the in vivo measurements.

5.2.1.3 Discussion

While the current model predicts muscle combinations which satisfy stability, these may not represent the actual loading pattern on the spine. Precise agreement between in vivo observations and the muscle combinations applied in the model is therefore not expected. However, comparison of in vivo observations and the requirement for force in the model in the erect and flex3 postures provides some validity to the model predictions.
Comparison with in vivo measurements of intra discal pressure provides some indication of the expected joint reaction for the muscle recruitment patterns applied in the model. Combinations of muscle groups which satisfied stability involved a maximum of six or seven muscle groups (flex2 and flex3 postures) and in most cases only two or three. Although these patterns are not indicative of the actual spinal loading, the predicted joint reactions for the model do not exceed the reported values in real life. Additional submaximal activation of other muscle groups, or alternate combinations close to in vivo recruitment patterns are likely to increase the joint reactions by a small amount, closer to the values calculated using the in vivo disc pressure measurements.
5.2.2 MUSCLE ROLES AND RECRUITMENT PATTERNS

The perpendicular distance between the thrustline and the centre line of the arch is proportional to the bending moments in the arch at that level (Heyman, 1982). The predicted thrustline due to body weight forces was projected anterior to the vertebral path and represents positive (anterior) bending moments within the arch. In contrast, the predicted thrustline due to the ‘curved’ representation of longissimus thoracis muscle forces was projected significantly posterior to the spinal configuration and represents negative (posterior) bending moments within the arch. Muscle groups which generate forces sufficient to significantly alter the bending moments within the arch, are recognised by their significant effects on the thrustline. Those which generate significant extension moments are longissimus thoracis, iliocostalis lumborum, multifidus and semispinalis capitis in the upright posture, whilst psoas and the abdominal muscles rectus abdominis and internal and external oblique generates significant flexion moments. These correspond to their respective roles of extension and flexion described in anatomic textbooks. The superficial lumbar erector spinae muscles longissimus thoracis and iliocostalis lumborum have also been reported to be well positioned to generate significant bending moments on the spine and control overall movement (Aspden, 1992, Bergmark, 1989).

The requirement for muscle force varies with each individual posture, according to the vertebral configuration and the overall direction of the reference line.
In the erect posture the amount of muscle force available was far greater than that required to counteract the flexion moment due to the external load and body weight in this posture. Rather the problem found was determining the fine combination of forces required to ensure the path of forces and the natural curvature of the spine coincide. This required local application of small amounts of muscle force in the cervical and lumbar regions. Electromyographic activity recorded for in vivo studies during upright standing confirms the inactivity of the large extensor muscles semispinalis capitis (Takebe et al, 1974) and the erector spine (Floyd and Silver, 1955).

In flexion, the force patterns identified in this model demonstrated the requirement for greater muscle force to be generated. The roles of the main extensor muscles (semispinalis capitis, longissimus thoracis and thoracic iliocostalis) became significant. Under these loading conditions, the effects of an increase in force due to the curved representation of longissmus thoracis and thoracic iliocostalis becomes evident. In the flex2 and flex3 postures maximal activation of these muscles is necessary to ensure stability in the lower levels of the spine when supporting an applied loading. This agrees with the observations of Schultz et al. (1985) who reported activation of the back muscles in the fully flexed posture when supporting an external load.
5.2.3 THE ROLE OF THE LIGAMENTS

Previously several authors have studied the contribution of the ligaments towards generating the required extensor moment to support the spine during lifting activities (e.g. McGill et al., 1986, Cholewicki and McGill, 1992). However, these studies reported the ligaments of the lumbar spine were not sufficiently stretched due to subjects maintaining a straight back (lumbar lordosis present) when lifting. The amount of force generated by the ligaments was therefore insufficient. Indeed McGill et al. (1986) reported the muscle forces to provide approximately 99% of the required extensor moment.

In this study, the ligaments were assumed to be sufficiently stretched to generate forces near to their maximum capability without injury. However, due to the close positioning of the points of application for each force pair, the ligaments were predicted to contribute little towards the overall stability of the spine. This model therefore implies the points of force application rather than force magnitude are limiting. Panjabi (1989) reported difficulty in separating the posterior longitudinal ligament from adjacent annulus fibres in the in vitro study performed. He implied the role of the ligament is to strengthen parts of the annulus that may be subjected to large stresses during flexion. Model predictions support this idea that ligaments and also intersegmental muscles are necessary for segmental stability rather than the whole spine.
5.2.4 MODEL PREDICTIONS AND EXISTING CONCEPTS

5.2.4.1 Muscle Coactivity

Coactivity of the abdominal muscles has been reported for both isometric (Marras et al., 1984) and dynamic lifting activities (Granata et al., 1995). Also McGill and Norman (1986) reported abdominal activity in the upright position following a lifting activity, in which the muscles were "attempting to stabilise the trunk during balancing of the external load and the weight of the upper body”.

Some studies have suggested the abdominal muscle forces are necessary to provide stiffness to the trunk, reducing the likelihood of instability. The orientation of the fascicles of these muscle forces relative to the L3/L4 joint axis has been reported by Dumas for the upright posture. Assuming these muscle forces maintain this orientation relative to the reference line of the spine with flexion, their direction can be compared with the orientation of forces due to body weight in each posture. The direction of the muscle forces relative to the vertical are reported in Table 5-8. (In the configuration used in this work the L3/L4 intervertebral joint axis is vertical).

Table 5-8: Reported Orientation of the Abdominal Muscle Fascicles by Dumas

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>Fascicle Direction relative to Vertical (degrees)</th>
<th>Fascicle Direction relative to Reference Line (degrees)</th>
</tr>
</thead>
<tbody>
<tr>
<td>External Oblique Fascicles</td>
<td>28.6, 27.0, 26.7, 24.5</td>
<td>27.36, 25.76, 25.46, 23.26</td>
</tr>
<tr>
<td>Internal Oblique Fascicles</td>
<td>36.3, 33.3, 33.0</td>
<td>35.06, 32.06, 31.76</td>
</tr>
<tr>
<td>Rectus Abdominis Fascicles</td>
<td>8.0, 9.3, 7.3, 11.5</td>
<td>6.76, 8.06, 6.06, 10.26</td>
</tr>
</tbody>
</table>
In the flex2 and flex3 postures, the orientation of the body weight forces increases to 49.51 and 74.72 degrees respectively relative to the reference line. The abdominal forces therefore have smaller negative perpendicular components in comparison with body weight and will have a flattening effect on the thrustline curve due to body weight at their points of application.

However, the predictions of the current model indicated the abdominal muscles increased the overall flexion moment on the spine when considered relative to body weight. Also when combined with the existing combination of muscles for which a good fit thrustline had been obtained in the flex2 posture, the overall thrustline curve was consequently increased and the spine was more unstable (Figure 20).

However, Granata et al. (1995) also reported that an increase in extensor muscle activity was needed to counteract the flexion moment due to the abdominal muscles. The consequent increase in compression due to the abdominal and extensor muscle activity was approximately 45% for dynamic lifting for a 36Kg load.
Additional muscle force due to lumbar iliocosalis and multifidus and forces due to intra abdominal pressure (16 Kpa) created through abdominal muscle contractions were therefore considered. The extensor muscle force was not sufficient to counteract the flexion moment of the abdominal muscles for the values used and slight instability occurred at the upper vertebral levels. For the current loading conditions there is therefore nothing apparent to be gained from the additional muscle force.

However, studies of stability of the lumbar spine have implied that muscle coactivity is necessary in order to provide sufficient stiffness to the spine to prevent buckling. The definition of stability used in these models is that upon a small disturbance the spine will return to the equilibrium position for which conditions of minimum potential energy are satisfied.
Therefore upon the application of an additional force in the model to create a small disturbance, stability can be assessed in terms of the position of the thrustline relative to the spine. Application of a 50N load perpendicular to the T8 vertebral surface for the combination of extensor and shoulder muscles which ensured stability resulted in a thrustline which moved outside the limits of the vertebral bodies at several levels (Figure 20). In contrast, application of the same load to the combination of muscles which included additional abdominal and extensor muscles resulted in a much smaller displacement of the thrustline (Figure 20).

The additional abdominal and lumbar muscle force therefore increases the magnitude of the compressive force in a direction which follows the path of the spine, so that application of a sudden load does not alter this direction sufficiently to move the thrustline outside the spine limits. When large loads are carried by the spine this may be vital towards ensuring stability and preventing incorrect loading of the spinal components.

5.2.4.2 Flexion -Relaxation Phenomenon

A number of studies have reported minimal activity of the large superficial spinal muscles in the fully flexed posture (Floyd and Silver, 1955, Schultz et al., 1985, Andersson et al., 1996). Schultz et al. (1985) report EMG activity measured at the L3/L4 level for this position was less than in the upright standing position and was minimal for both surface and deep erector spinae muscle activity.
Floyd and Silver (1955) suggested in this position the ligaments are the dominant forces acting to support the spine, while Adams et al., (1980 op cit. Adams and Dolan, 1996) have reported in full flexion the ligaments resist approximately 70% of the bending moment. In contrast, Andersson et al. (1996) reported when the trunk was kyphotic the quadratus lumborum and lateral deep erector spinae became activated.

The largest fascicle of the muscle longissimus thoracis, which is one of the main extensor muscles of the spine, can generate a force of approximately 178N. In comparison a single fascicle of the ligaments can generate a maximum of 1196.24N. However, the amount of force generated within the ligaments depends upon the degree of stretching and hence the degree of spinal flexion.

In the current study the flex3 posture was considered close to the full range of spinal flexion. However, model predictions indicate although the ligament forces may be considerably large when stretched, the close proximity of their points of attachment results in only a small flattening effect on the spine. When combined with the intersegmental muscles (interspinalis, intertransverse and multifidus) a further flattening effect was observed but this was not sufficient to ensure stability. Additional force is therefore required in this posture, supporting the findings of Andersson et al. (1996).
The additional force may be due to muscle passive elasticity. The applied force in this model is assumed to be generated by active contractile elements of the muscles and the passive force generated by stretching of the muscle with flexion has been ignored. McGill and Norman (1986) previously estimated the contribution of the lumbar muscle forces due to stretching to the overall extension moment required for a lifting activity. They reported due to the lifting strategy employed in which subjects maintained lumbar lordosis, the contribution due to muscle stretching was approximately less than 2%. However, fascicles of longissimus thoracis and thoracic iliocostalis which extend over several levels of the thoracic spine were predicted in this model to be significantly affected by curvature in the flexed postures. The length of these fascicles is therefore likely to be increased by the curvature and the passive components of force may be more significant.

5.2.4.3 Intra abdominal pressure

The transverse abdominis muscle has been observed to be active throughout dynamic lifting at an approximate constant level of contraction (Cresswell et al., 1992). The authors proposed this muscle is responsible for generating intra abdominal pressure. The direction of both the muscle and intra abdominal pressure forces almost perpendicular to the intervertebral joints allows the external loads to be resisted with minimum compressive penalty. The model predictions indicated both the transverse abdominis muscle and intra abdominal pressure to result in a flattening of the thrustline in the lumbar regions in all postures. In the upright posture the forces due to intra abdominal pressure in combination with muscle forces was predicted to satisfy conditions of stability for the whole spine.
This may be beneficial when ensuring the path of the thrustline corresponds to the path of the vertebral bodies in the erect posture and in flexion when lumbar lordosis is present. These forces may also be required to offset the increased curvature of the thrustline with abdominal muscle coactivity. Further detailed study is required in this area to establish the precise roles.

5.3 MODEL LIMITATIONS AND DEVELOPMENTS

5.3.1 WHOLE SPINE CONFIGURATION

The precise range of motion of the spine depends upon vertebral and disc dimensions and their stiffness properties. Variation at each level, and for each individual ensures the overall movement is unique for an individual. Also individuals perform lifting tasks with different strategies. The precise posture for a given individual performing a lifting task can therefore not be determined without in vivo records. This may explain the lack of published data regarding a description of the vertebral configuration for the whole spine.

Four postures were therefore simulated and were considered to be representative of the various stages during a lifting activity. Using measured vertebral and disc dimensions, the relative angles of orientation were approximated to correspond to anatomy and ergonomic textbook pictorial configurations. A comparison with the overall range of flexion-extension of each vertebra ensured these postures were within theoretical limits.
A study by Snijders and van Riel (1987) describes a non invasive technique which determines changes in spinal curvature skin profiles. Further study of the configuration of the spine in various postures and whilst performing lifting tasks using such non invasive techniques would provide a more accurate representation of the spinal curvature in vivo.

5.3.2 APPLICATION OF THE EXTERNAL LOAD

In the current study, the external load carried was assumed to be a direct extension of the weight of the arms, and was applied as an additional load at T2, T3 and T4 levels. Flattening of the thrustline curve mainly by lumbar muscle activity was not sufficient to result in stability at the upper levels of the spine in the flex2 and flex3 postures. However the steepness of the thrustline curve at these levels was attributed to the direct increase in force at these levels.

Muscles attaching between the spine and the shoulder complex have attachments distributed along the cervical and thoracic levels. Forces carried in the arms could be therefore transferred from the shoulders to the spine at several thoracic levels through this muscle activity. The effects of the shoulder muscles therefore need to be considered in order to distribute the external load more evenly along the spine. Detailed studies performed by Johnson et al. (1994) report the precise attachments and cross sectional areas of the trapezius and rhomboid muscles which may benefit this study.
Also in this study the spine was considered to be near-maximally challenged by the lifting of a 900N external load. This load was therefore assumed to be held close to the spine. However, consideration of other lighter loads held at a certain distance from the spine may be representative of tasks performed in industry in confined workspaces. Changes in the eccentricity of the load with arm position must therefore be considered for such investigations in the future.

5.3.3 MUSCLE AND LIGAMENT FORCE DISTRIBUTION PATTERNS

Lack of data in the literature required assumptions to be made about the distribution and magnitude of forces in the cervical and thoracic region. Force distribution patterns within the lumbar region muscles did not show any distinct trend upon which to base these assumptions. A uniform force distribution pattern was therefore assumed whereby the total cross sectional area reported in the literature represents the total area of a number of fascicles in that muscle group of approximately equal cross section. The sensitivity analysis performed showed distribution patterns which increased or decreased descending the spine had less than a 2.5% decrease on the overall depth of the thrustline curve (values for multifidus were higher). Slightly larger effects were observed for the intersegmental muscles with an increase in force magnitude. However when these effects were considered in combination, these muscle forces were insufficient to ensure stability.
Also for ligamentum flavum, the data reported by Chazal et al. (1985) did not include the cervical levels, even though anatomic textbooks describe this ligament to act along the whole spine (Gray's anatomy). In this model values were therefore estimated for these levels assuming the ligament in the cervical region is smaller than other regions. Similarly for some muscle groups such as ilio-costalis cervicis muscle strengths were estimated based upon their comparative size with other muscles.

So that a complete set of data for the ligaments and muscles could be obtained, values for the muscles and ligaments included in the model were of various sources, and various degrees of detail. To overcome this limitation, the values for muscle and ligament forces have been parameterised so that data generated in future anatomic studies may be readily applied as input data to the model.

### 5.3.4 INTRA ABDOMINAL PRESSURE

In this study forces due to intra abdominal pressure were applied based upon values reported in the literature. However realistically, intra abdominal pressure is dependent upon posture and force values vary with vertebral configuration. A more accurate investigation would therefore require data for spinal configuration and corresponding intra abdominal pressure measurements.

Also, in this study the combined effects of intra abdominal pressure and the abdominal muscle forces necessary to generate this pressure were not considered.
5.3.5 CURVED MUSCLE REPRESENTATION

Studies in the literature describing changes in muscle lines of action with flexion are few (Aspden 1992). One model of the cervical spine developed by Deng and Goldsmith (1987) considered motion of the spine in response to impact. This required consideration of changes in spinal curvature, and therefore changes in muscle force direction. In the erect posture the muscle forces were assumed to act in a straight line between their attachment points. However an intermediate point was also defined through which these forces were required to act. In the erect posture, this point was midway between the attachment points. In flexion, this point would ensure the path of the muscle was maintained relative to the spinal configuration.

In this study the method used to represent muscle curvature considered the forces directed to the peak of the spinal curve between their attachment points. This was considered a more accurate representation which could detect for the most curved part of the spine spanned by a muscle, not necessarily at the midpoint of the muscle or the midpoint of the spinal region spanned.

However as considered in the sensitivity analysis, this method could not correct for the muscle longissimus thoracis in the erect and flex1 postures. Further development of this algorithm could therefore be pursued.
5.3.6 DYNAMIC INVESTIGATIONS

To perform a lifting activity muscle force must be generated to overcome the inertia of the trunk and initiate motion. The muscle force required to perform a dynamic lifting activity is greater than that required for supporting the spine in a single position.

In the current model, instability predicted at the upper thoracic levels in the flex2 and flex3 posture when supporting an external load was attributed to the absence of the shoulder muscles. The three major extensor muscle groups longissimus thoracis, thoracic iliocostalis and semispinalis capitis were sufficient to ensure stability in the lower levels of the spine, and additional lumbar muscle force resulted in instability. The muscle force was therefore sufficient to support the spine in a single position. However under dynamic loading conditions, additional muscle force may be needed.

Marras and Mirka (1996) have reported velocity and acceleration associated with dynamic lifting activities to be more sensitive measures of low back pain than the range of motion. Dynamic loading conditions may therefore indicate spinal regions in which muscle force is deficient. However before dynamic loading can be considered, the roles of muscle coactivity and shoulder muscles in the current model must be clarified.
5.4 APPLICATION OF THE MODEL TO FUTURE LOW BACK PAIN INVESTIGATIONS

The current model applies a distributed loading pattern for muscle forces which allows the role of the individual muscle groups to be considered. The demands on a particular muscle group in contributing to stability can therefore be assessed for a given posture. The effects of a reduction in strength of this group, through fatigue, injury, or through inactivity due to sedentary work on stability of the spine can be appreciated.

Consideration of the configuration of the whole spine allows stability at every level to be considered. The effects of changes in loading or in curvature in a certain region on other regions of the spine can therefore be investigated. Postures can be assessed for their demands on muscle activity at every level, and for possible causes of pain cited in the literature at neck, shoulder and low back levels.

The application of distributed forces due to body weight and muscles allows prediction of the loading at each intervertebral joint. The current model indicates the maximum level of loading varies with posture and muscle forces, and is not necessarily associated at the lowest lumbar level. Possible causes of back pain at levels other than the lumbar regions can again be considered.
The model predictions indicate muscles with the greatest effects on the overall thrustline, and those with localised effects in the cervical, thoracic and lumbar levels. Selection of a muscle group can be made upon the predictions of the current study when testing other postures and lifting tasks. With the recommended developments, including a better distribution of the external load along the spine through shoulder muscle activity, the model can therefore be an invaluable tool in investigating the stability and loading of the spine.

The current model predictions indicate the muscle forces are capable of supporting the upper body weight in the postures tested, and are capable of supporting an external load of 900N with additional activation of the shoulder muscles. Considerable biological variation exists between individuals for spinal dimensions and material strength, muscle forces and recruitment, and body weight. Each individual therefore adopts a unique lifting strategy with the minimal demand on the muscles and joints when performing the task in their own preferred way. However constraints imposed upon the spine such as the magnitude of the load to be lifted, or the space in which the lift can be performed may restrict the curvature of the spine, and may place greater demands upon the muscle forces to ensure stability.

Thus while each individual may perform a lifting activity successfully by adopting a preferred lifting strategy, the individual may not be comfortable when the method of lifting is restricted. Additional biological constraints include a reduction in muscle strength through injury or inactivity, an increase in body weight or a change in spinal curvature associated with pregnancy. Also work place restrictions such as the amount of space in which to perform the task, the size of the object to be lifted, or the design of equipment may limit the method of performance.
Future investigations using the current mathematical model as a tool for predicting spinal stability and muscle and joint loading may therefore be directed towards constraints imposed by task or workplace requirements, or biological variation.
CHAPTER SIX:

6. CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE RESEARCH

6.1 BASIC CONCLUSIONS

Current guidelines (e.g. NIOSH in the US and EC Directive in the UK) regarding lifting activities in industry are based upon the predictions of the lever model. This considers the amount of muscle force required to satisfy equilibrium of the spine about a single intervertebral joint. The predicted compression and muscle forces are used to determine the loads which should be lifted.

However, Aspden (1987) also proposed that the curvature of the spine can have a significant load bearing role. When representing the spine as an arch, the model predictions favoured a significant reduction in muscle force required and the consequent loading at the L5/S1 joint.

More recent models also indicate equilibrium of the spine does not necessarily imply stability, and that stability may be associated with tissue injury and low back pain.

Aspden, 1987 reported that the spine is stable in a given posture when the applied loads can be transmitted between vertebrae along the entire length of the spine. This depends upon the curvature. Loading of the muscles and intervertebral joints can therefore be influenced by posture.
However, investigations regarding lifting activities which consider spinal curvature are limited to those performed by Aspden, which included collective loading for the muscle forces and for body weight, and only partial representation of the spine. On the basis that the level of anatomic detail of a model affects the accuracy of its predictions (McGill and Norman, 1987), a model which included greater anatomic detail based upon arch theory was developed.

Particular features considered necessary were:
1) Whole spinal curve
2) Body weight forces applied at each vertebral level
3) Individual muscle groups

The effects of these features were considered for four postures at representative stages of a lifting activity.

1) Whole spinal curve

The investigations performed indicated several cases for which stability of spine had been satisfied at some, but not all vertebral levels. Stability of the lumbar spine which has been the focus of several previous models does therefore not imply stability of the whole spine.

Additional muscle force to satisfy stability at every level affects the consequent joint reactions. In particular, addition of the shoulder muscles in the flex3 posture required to support an external load of 900N resulted in an increase in joint reactions at the thoracic and lumbar vertebral levels. Consideration of the whole spine is therefore necessary to determine loading for a particular region or level.
The requirements for muscle force at each level, and the consequent joint reactions can then be considered for their contribution to neck, shoulder and low back pain cited in the literature.

2) Body weight forces applied at each vertebral level

Consideration of body weight at each vertebral level was viewed necessary to provide a loading pattern which matched the precise curvature of the spine. Comparison with the collective loading pattern showed a smoother thrustline curve.

Consideration of the eccentricity of each body weight force was also found to be significant. In the erect posture, the collective loading was found to under estimate the amount of muscle force required to support the weight of the upper body and the external load.

Predicted joint reactions also indicated that the distributed loading pattern is necessary for consideration of the loads at each vertebral level. A more gradual change in force was observed in comparison with the abrupt changes in shear and compression with the application of collective loads.

3) Individual muscle groups

Consideration of the forces due to body weight alone was considered essential to determine the amount of muscle force required for stability in each posture.

The study of each individual muscle group when added to body weight forces was an initial step towards identifying muscles which contribute significantly to the stability of the whole spine. Based on this combinations of muscle groups which ensured stability of each posture were determined.
The study revealed muscle groups in the cervical and lumbar regions which are necessary for local control of the thrustline. The thrustline could be made to fit the lordotic curves of the cervical and lumbar vertebrae which could not be achieved with a collective muscle force. Larger muscle groups were more significant in the flexed postures when supporting an external load. However, in the flex2 and flex3 postures the study indicated a local requirement for force in the thoracic region. Further testing revealed activation of the shoulder muscles satisfies this requirement. Again the role for individual muscle groups could not be detected using the collective loading.

The model predicted individual muscle groups acting over certain levels of the spine had a smaller effect than the collective force acting over the whole length. Consequently, predicted compression at the intervertebral joints was greater. However, these values were considerably less than those predicted for a lever model.

Application of individual muscle groups also resulted in an increase in compression at the intervertebral joint other than the lumbar region. In particular, activation of the semispinalis capitis increased the loading at cervical levels, while the activation of the shoulder muscles considerably increased the compression at the thoracic levels.
In comparison with the collective loading the developed model allows a number of possible investigations:

1) The effects of a change in configuration at a certain level of the spine on the muscle forces and joint loading can be considered through the whole spine model.

2) The demands on a particular muscle group in contributing to stability can be assessed for a given posture. Association between posture, muscle injury and back pain can therefore be investigated.

3) By reducing the amount of muscle force applied, the effects of a reduction in strength of a muscle group, through fatigue, injury, or through inactivity due to sedentary work on the stability of the spine can be appreciated.

4) The loads at a single or many intervertebral joints due to a change in activation level of a particular muscle group can be determined.

These features may be more significant when investigating possible causes of back pain in any region.

The model predictions showed combinations of between two and six muscle groups were sufficient to ensure conditions of stability of the spine for each posture. A number of other possible thrustlines can be obtained with alternative recruitment patterns of muscles in various states of contraction. However, based upon the proposal by Aspden, the existence of at least one thrustline which satisfies stability was used to indicate the spine can be supported.
In the current study, the spinal muscles were predicted to be sufficient to support the weight of the upper body in the four postures representative of various stages of trunk flexion. For these postures the predicted compression at each intervertebral joint was well below the known failure values of the vertebral end plate, the weakest link in the motion segment. It is therefore unlikely that when performing this activity intervertebral compression is a limiting factor.

In flexion, the application of a 900N load was considered to present a near-maximal challenge. An additional extensor force available in the lumbar regions due to multifidus and iliocostalis muscles and intra abdominal pressure was not required. However, activation of the thoracic muscles to ensure stability at these levels was essential in the flex2 and flex3 postures. The initial assumption that the shoulder muscles contribute little to spine stability therefore does not hold.

The predicted compression at the lumbar intervertebral joints was again below the known failure values for the vertebral end plates. Although slight instability remained in the flex3 posture, submaximal activation of additional shoulder muscles to overcome this are unlikely to increase the joint compression to failure values. Thus, when lifting loads of lower magnitude as defined by industrial guidelines, the amount of muscle force and the joint compression values are unlikely to be limiting factors.
6.2 RECOMMENDATIONS FOR FUTURE RESEARCH

This study has assumed static loading conditions. Greater muscle force is required to overcome the inertia forces of the trunk in dynamic lifting. Also loads are assumed to be held close to the chest. Work tasks which require loads to be held or lifted further from body (typical of nursing for example), require greater muscle force to balance the increased flexion moments. Repetition of these tasks may also lead to muscle fatigue. Finally, work place restrictions such as the amount of space in which to perform the task, the size of the object to be lifted, or the design of equipment may limit the choice of lifting strategy and place greater demands upon the muscle forces to ensure stability. For possible causes of low back pain, task related factors must therefore be investigated.

Biological factors such as an increase in body weight, different body weight distribution, variation in muscle strength, or curvature must also be investigated. A work task performed by a certain individual may be unsuitable for a pregnant woman who undergoes changes in all these features.

The amount of data available in the literature regarding spinal configuration and muscle force properties is limiting to the current investigations. The current study investigates stability of four postures for a single set of vertebral dimensions and muscle forces. However, parametrization of these features within the model will allow the model to be readily included in future investigations for a wide range of postures and loading conditions. The model can then be used to test for biological variation within the population, or for the spine of a specific individual.
Previous models which investigate stability have considered the lumbar spine (e.g. Bergmark et al., 1989) or the thoraco-lumbar spine (e.g. Shirazi Adl, 1993).

However, the current study shows consideration of the whole spine is necessary. Also these are based upon stability using Euler theory whereby the spine is represented as an elastic column. This takes into account the tensile properties of the disc in bending. In contrast, the existing model considers the compressive forces transferred between vertebrae and is based upon the safe theory of arches. This considers postures and loading conditions for which the forces are transmitted between vertebral bodies, thereby alleviating the loads supported by the tensile properties of the disc and joint ligaments. By considering 'safe' postures with this model, possible causes of back pain through excess loads on the disc and ligaments can therefore be reduced.
7. REFERENCES


op cit.


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7.1 BIBLIOGRAPHY

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8. APPENDIX

8.1 VALIDATION OF THE RELATIVE ANGLES USED FOR VERTEBRAL CONFIGURATIONS

Table 8-1: Comparing the relative angles used for the flexed postures with the overall range of flexion and extension at each vertebral level.

<table>
<thead>
<tr>
<th>Vertebra</th>
<th>Relative angles. Erect posture</th>
<th>Relative angles. Flex1 posture</th>
<th>Relative angles. Flex2 posture</th>
<th>Relative angles. Flex3 posture</th>
<th>Range of flexion-extension /degrees</th>
</tr>
</thead>
<tbody>
<tr>
<td>C1/C2</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>10</td>
</tr>
<tr>
<td>C2/C3</td>
<td>-3</td>
<td>6.5</td>
<td>5</td>
<td>5.5</td>
<td>8</td>
</tr>
<tr>
<td>C3/C4</td>
<td>3</td>
<td>3.5</td>
<td>5</td>
<td>5</td>
<td>13</td>
</tr>
<tr>
<td>C4/C5</td>
<td>6</td>
<td>2</td>
<td>-2</td>
<td>7</td>
<td>12</td>
</tr>
<tr>
<td>C5/C6</td>
<td>7</td>
<td>1</td>
<td>-2</td>
<td>5</td>
<td>17</td>
</tr>
<tr>
<td>C6/C7</td>
<td>6</td>
<td>0.5</td>
<td>-4</td>
<td>3</td>
<td>16</td>
</tr>
<tr>
<td>C7/T1</td>
<td>4</td>
<td>1.5</td>
<td>0.5</td>
<td>1</td>
<td>9</td>
</tr>
<tr>
<td>T1/T2</td>
<td>2</td>
<td>-2.5</td>
<td>-3</td>
<td>-5</td>
<td>4</td>
</tr>
<tr>
<td>T2/T3</td>
<td>-6</td>
<td>-2</td>
<td>-5</td>
<td>-1.5</td>
<td>4</td>
</tr>
<tr>
<td>T3/T4</td>
<td>-4</td>
<td>-3.5</td>
<td>-2.5</td>
<td>-2</td>
<td>4</td>
</tr>
<tr>
<td>T4/T5</td>
<td>-1</td>
<td>-4.5</td>
<td><strong>-7.5</strong></td>
<td>-2.5</td>
<td>4</td>
</tr>
<tr>
<td>T5/T6</td>
<td>-7</td>
<td>-6.5</td>
<td>-7</td>
<td>-7</td>
<td>4</td>
</tr>
<tr>
<td>T6/T7</td>
<td>-1</td>
<td>-5.5</td>
<td>-5</td>
<td>-6</td>
<td>5</td>
</tr>
<tr>
<td>T7/T8</td>
<td>-3</td>
<td>-4.5</td>
<td>-3</td>
<td>-5</td>
<td>6</td>
</tr>
<tr>
<td>T8/T9</td>
<td>-4</td>
<td>-3.5</td>
<td>-7</td>
<td>-5</td>
<td>6</td>
</tr>
<tr>
<td>T9/T10</td>
<td>-4</td>
<td>-2.5</td>
<td>-4.5</td>
<td>-5</td>
<td>6</td>
</tr>
<tr>
<td>T10/T11</td>
<td>-2</td>
<td>-1.5</td>
<td>-3</td>
<td>-5</td>
<td>9</td>
</tr>
<tr>
<td>T11/T12</td>
<td>-1</td>
<td>-0.5</td>
<td>-3</td>
<td>-6</td>
<td>12</td>
</tr>
<tr>
<td>T12/L1</td>
<td>2</td>
<td>2</td>
<td>-5</td>
<td>-5</td>
<td>12</td>
</tr>
<tr>
<td>L1/L2</td>
<td>5</td>
<td>3.5</td>
<td>-1</td>
<td>-4</td>
<td>12</td>
</tr>
<tr>
<td>L2/L3</td>
<td>8</td>
<td>5</td>
<td>0</td>
<td>-3</td>
<td>14</td>
</tr>
<tr>
<td>L3/L4</td>
<td>11</td>
<td>12</td>
<td>7</td>
<td>-1</td>
<td>15</td>
</tr>
<tr>
<td>L4/L5</td>
<td>14</td>
<td>10.5</td>
<td>11</td>
<td>5</td>
<td>16</td>
</tr>
<tr>
<td>L5/S1</td>
<td>15</td>
<td>7</td>
<td>15</td>
<td>15</td>
<td>17</td>
</tr>
<tr>
<td>Sacrum relative to global y coordinate axis</td>
<td>-40</td>
<td>-44</td>
<td>-55</td>
<td>-65</td>
<td></td>
</tr>
</tbody>
</table>

The value shown in bold print indicates the relative angle used is greater than the known flexion-extension range. However this can be corrected for by updating the appropriate parameter when more accurate data becomes available.
8.2 REPRESENTATION OF THE POINTS OF MUSCLE ATTACHMENT TO THE VERTEBRAE

The position of each representative point for muscle attachment on the spinous process, transverse process, superior articulating processes and vertebral lamina was required. For each vertebral level, an expression defining the x and y co-ordinates of these points relative to the origin of the vertebral body was used. Parameter values reported by Stepney et al. (1996) were then used to derive the co-ordinate values for each vertebral level.

8.2.1 CALCULATION OF LOCATION OF SPINOUS PROCESS RELATIVE TO VERTEBRAL ORIGIN

\[ x \text{ co-ordinate: } \frac{1}{2} \ast \text{vblength} + \text{sclength} + \text{splength} \ast \cos(\text{spit}) \]

\[ y \text{ co-ordinate: } \text{vbheight} - \text{splength} \ast \sin(\text{spit}) \]

These parameters are illustrated in Figure 8-1 where:

\[
\begin{align*}
\text{vblength} & = \text{length of vertebral body} \\
\text{sclength} & = \text{length of spinal canal} \\
\text{splength} & = \text{length of spinous process} \\
\text{spit} & = \text{angle of spinous process relative to the horizontal}
\end{align*}
\]

\[\]

\[\]

\[1\] Although the top of the spinous process is not directly level with the height parameter measure of the vertebral body, this was assumed to be a close approximation for the y co-ordinate.
8.2.2 CALCULATION OF LOCATION OF TRANSVERSE PROCESS RELATIVE TO VERTEBRAL ORIGIN

The position of the representative point on the transverse processes in the sagittal plane depends upon dimensions and orientation of both the pedicles and transverse processes and the in the lateral and sagittal planes.

\[ x \text{ co-ordinate: } \frac{v\text{length}}{2} + \text{pedicle x distance} + \text{transverse process x distance} \]

This is illustrated in Figure 8-2.
Figure 8-2: Illustration of the position of the transverse process representative point relative to the origin of the vertebral body

Figure 8-3 provides a simple illustration of the inclination of the pedicles relative to the vertebral body. The x distance of the end point of the pedicles relative to the vertebral body origin is:

\[(v\text{length}/2) + \text{inclined distance} \times \cos(\text{inclination angle})\]  

(1)
However in the plane of the pedicle, the inclined distance is affected by the lateral inclination of the pedicles. In Figure 8-4 the inclined distance is:

\[ \text{pedlength} \times \cos(\text{pdit}) \] (2)

![Figure 8-4: Illustration of the inclination of the pedicles in the lateral plane.](image)

Returning to the plane of the vertebral origin, the x distance of the end point of the pedicles relative to the vertebral body origin defined by (1) and (2) is:

\[ \frac{\text{vblength}}{2} + (\text{pedlength} \times \cos(\text{pdit})) \times \cos(\text{pdis}). \]

In the sagittal plane the y co-ordinate of the end of the pedicle is unaffected by lateral orientation, and is \( \text{pedlength} \times \sin(\text{pdit}) \).

Similarly the x co-ordinate of the transverse processes is affected by the orientation in both the sagittal (tpil) and lateral planes (tpis), whereas the y co-ordinate is only determined by the sagittal angle.

The x distance of the tip of the transverse process relative to the end point of the pedicles is:

\[ (\text{tpleNGTH} \times \cos(\text{tpil})) \times \cos(\text{tpis}). \]
The x co-ordinate of the representative point on the transverse process is therefore:
\[ vbheight + \text{pedicle y distance} + \text{transverse process y distance} \]

\[ = \left[ \left( \frac{vblength}{2} \right) + \left( \text{pedlength} \times \cos (\text{pdit}) \right) \times \cos (\text{pdis}) \right] + \left( \text{tplength} \times \cos (\text{tpil}) \right) \times \cos (\text{tpis}) \] 

The y co-ordinate of the representative point on the transverse process is therefore:
\[ vbheight + \text{pedicle y distance} + \text{transverse process y distance} \]

\[ = [vbheight + (\text{pedlength} \times \sin (\text{pdit})) + (\text{tplength} \times \sin (\text{tpil}))] \]

8.2.3 CALCULATION OF LOCATION OF SUPERIOR ARTICULAR PROCESS IN THE CERVICAL REGION RELATIVE TO VERTEBRAL ORIGIN

x co-ordinate: \( \frac{1}{2} \times vblength + sfl \times \cos (sfit) \)

y co-ordinate: \( vbheight + sfl \times \sin (sfit) \). These are illustrated in Figure 8-5.

Figure 8-5: Illustration of the position of the superior articulating surface representative point relative to the origin of the vertebral body (shown for thoracic but parameters are the same).
8.2.4 Calculation of location of mamillary processes in the lumbar region relative to vertebral origin

\[ \text{x co-ordinate: } \frac{3}{2} \times \text{vblength, } \quad \text{y co-ordinate: } \text{vbheight} + \frac{1}{2} \times \text{sfl} \times \sin(\text{sfit}). \]

These are illustrated in Figure 8-6.

**Figure 8-6:** Illustration of the position of the mamillary process relative to the origin of the vertebral body.
8.2.5 CALCULATION OF LOCATION OF VERTEBRAL LAMINA IN THE THORACIC REGION RELATIVE TO VERTEBRAL ORIGIN

x co-ordinate: \( \frac{vblength}{2} + sclength \),
y co-ordinate: \( \frac{vbheight}{2} \).

This is illustrated in Figure 8-7.

Figure 8-7: Illustration of the position of the vertebral lamina relative to the origin of the vertebral body.
8.3 CALCULATION OF CO-ORDINATES OF POINTS OF ATTACHMENT OF
MUSCLE GROUPS TO THE SKULL

The precise descriptions of the neck muscle points of attachment to the skull were
obtained from anatomic textbooks (see page 208) and are reported in Table 8-2.

Table 8-2: Points of neck muscle attachment to the head.

<table>
<thead>
<tr>
<th>Muscle name</th>
<th>External attachment point</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus capitis anterior</td>
<td>base of occipital bone in front of foramen magnum</td>
</tr>
<tr>
<td>Rectus capitis lateralis</td>
<td>inferior surface of jugular process of occipital bone</td>
</tr>
<tr>
<td>Rectus capitis posterior major</td>
<td>lateral part of inferior nuchal line of occipital bone</td>
</tr>
<tr>
<td>Rectus capitis posterior minor</td>
<td>occipital bone below inferior nuchal line</td>
</tr>
<tr>
<td>Splenius capitis</td>
<td>mastoid process of temporal bone and lateral part of superior nuchal line</td>
</tr>
<tr>
<td>Spinalis capitis</td>
<td>occipital bone between superior and inferior nuchal lines, between semispinalis capitis posterior minor (Martin 1994)</td>
</tr>
<tr>
<td>Semispinalis capitis</td>
<td>medial impression between superior and inferior nuchal lines of occipital bone</td>
</tr>
<tr>
<td>Longus capitis</td>
<td>inferior surface of basilar part of occipital bone</td>
</tr>
<tr>
<td>Longissimus capitis</td>
<td>posterior margin of the mastoid process</td>
</tr>
<tr>
<td>Obliquus capitis superior</td>
<td>inferior nuchal line of occipital bone</td>
</tr>
</tbody>
</table>
8.3.1 X CO-ORDINATES

The illustrations used from Gray's anatomy (see page 216) provided the relative positions of the muscle attachment points to the skull in the sagittal plane at the level of the occipital condyles. The centre of the foramen magnum was assumed to correspond to the centre of the neural canal of the atlas vertebra directly below. The distances of each muscle attachment point relative to the centre of the atlas vertebra were therefore obtained. Using scaled cross sectional anatomy drawings (Eccleshymer and Shoemaker, 1911), measurements were then scaled according to the length of the skull in the sagittal plane at this level. These were then expressed relative to the C1 origin in the model by their position relative to the centre of the dens (assumed to coincide with the C1 origin in the sagittal plane). Finally, these were scaled by a factor of $5/4$ to correspond to actual size measurements:
8.3.2 Y CO-ORDINATES

Illustrations in Grays anatomy (page 217) were used to identify the relative positions of the muscle points of attachment in vertical plane and the superior surface of that atlas. These distances were scaled by a factor of two to agree with the dimensions in the cross sectional anatomy drawings for the distance from the superior surface of the atlas to the top of the head, and by a factor of $5/4$ to produce actual size dimensions. To express these relative to the origin of the Cl vertebra required the addition of the Cl vertebral height to these values.

The approximate $x$ and $y$ co-ordinates are shown in Table 8-3.

Table 8-3: Calculated $x$ and $y$ co-ordinates for muscle attachment points to skull.

<table>
<thead>
<tr>
<th>Muscle name</th>
<th>$X$ co-ordinate relative to Cl origin (mm)</th>
<th>$Y$ co-ordinate relative to superior surface of atlas (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus capitis anterior</td>
<td>-5.57</td>
<td>12.5</td>
</tr>
<tr>
<td>Rectus capitis lateralis</td>
<td>3.52</td>
<td>12.5</td>
</tr>
<tr>
<td>Rectus capitis posterior major</td>
<td>29.66</td>
<td>31.25</td>
</tr>
<tr>
<td>Rectus capitis posterior minor</td>
<td>37.61</td>
<td>31.25</td>
</tr>
<tr>
<td>Splenius capitis</td>
<td>26.25</td>
<td>12.5</td>
</tr>
<tr>
<td>Spinalis capitis</td>
<td>54.66</td>
<td>35.94</td>
</tr>
<tr>
<td>Semispinalis capitis</td>
<td>54.66</td>
<td>40.625</td>
</tr>
<tr>
<td>Longus capitis</td>
<td>-11.25</td>
<td>12.5</td>
</tr>
<tr>
<td>Longissimus capitis</td>
<td>20.57</td>
<td>12.5</td>
</tr>
<tr>
<td>Obliquus capitis superior</td>
<td>31.93</td>
<td>31.25</td>
</tr>
</tbody>
</table>
8.4 THRUSTLINE CATEGORISATION

In order to compare the relative effects of each muscle group, muscles were included in categories which best described their effect on the predicted thrustline. These were:

Category One: Thrustlines moved anterior to the reference line.
Category Two: Thrustlines moderately flattened.
Category Three: Thrustlines moderately flattened and localised effect in cervical region.
Category Four: Thrustlines flattened by small amount and localised effect in cervical region.
Category Five: Thrustlines flattened by small amount.
Category Six: Negligible Effect on overall thrustline.
Category Seven: Thrustlines moderately flattened and localised effect in lumbar region.
Category Eight: Thrustlines curvature moderately increased.
Category Nine: Thrustlines curvature significantly increased.

The predicted thrustlines for the muscles in each category are presented here for the flexed postures, and for categories four and five in the erect posture which were not shown in chapter 4. For the flex2 posture no muscles were included in category 4.
8.4.1 ERECT POSTURE

Figure 8-8: The predicted thrustlines for category 4 muscle forces in the erect posture combined with body weight.

Figure 8-9: The predicted thrustlines for category 5 muscle forces in the erect posture combined with body weight.
8.4.2 FLEX1 POSTURE

Figure 8-10: The predicted thrustlines for category 1 muscle forces in the flex1 posture combined with body weight.

Figure 8-11: The predicted thrustlines for category 2 muscle forces in the flex1 posture combined with body weight.
Figure 8-12: The predicted thrustlines for category 3 muscle forces in the flex1 posture combined with body weight.

Figure 8-13: The predicted thrustlines for category 4 muscle forces in the flex1 posture combined with body weight.
Figure 8-14: The predicted thrustlines for category 5 muscle forces in the flex1 posture combined with body weight.

Figure 8-15: The predicted thrustlines for category 6 muscle forces in the flex1 posture combined with body weight.
Figure 8-16: The predicted thrustlines for category 7 muscle forces in the flex1 posture combined with body weight.

Figure 8-17: The predicted thrustlines for category 8 muscle forces in the flex1 posture combined with body weight.
Figure 8-18: The predicted thrustlines for category 9 muscle forces in the flex1 posture combined with body weight.

8.4.3 FLEX2 POSTURE

Figure 8-19: The predicted thrustlines for category 1 muscle forces in the flex2 posture combined with body weight.
Figure 8-20: The predicted thrustlines for category 2 muscle forces in the flex2 posture combined with body weight.

Figure 8-21: The predicted thrustlines for category 3 muscle forces in the flex2 posture combined with body weight.
Figure 8-22: The predicted thrustlines for category 5 muscle forces in the flex2 posture combined with body weight.

Figure 8-23: The predicted thrustlines for category 6 muscle forces in the flex2 posture combined with body weight.
Figure 8-24: The predicted thrustlines for category 7 muscle forces in the flex2 posture combined with body weight.

Figure 8-25: The predicted thrustlines for category 8 muscle forces in the flex2 posture combined with body weight.
Figure 8-26: The predicted thrustlines for category 9 muscle forces in the flex2 posture combined with body weight.
8.4.4 FLEX3 POSTURE

Figure 8-27: The predicted thrustlines for category 1 muscle forces in the flex3 posture combined with body weight.

Figure 8-28: The predicted thrustlines for category 2 muscle forces in the flex3 posture combined with body weight.
Figure 8-29: The predicted thrustlines for category 3 muscle forces in the flex3 posture combined with body weight.

Figure 8-30: The predicted thrustlines for category 5 muscle forces in the flex3 posture combined with body weight.
Figure 8-31: The predicted thrustlines for category 6 muscle forces in the flex3 posture combined with body weight.

Figure 8-32: The predicted thrustlines for category 7 muscle forces in the flex3 posture combined with body weight.
Figure 8-33: The predicted thrustlines for category 8 muscle forces in the flex3 posture combined with body weight.

Figure 8-34: The predicted thrustlines for category 9 muscle forces in the flex3 posture combined with body weight.
8.5 ILLUSTRATIONS OF THE LINES OF ACTION OF MUSCLE GROUPS REFERED TO IN THE TEXT

Figure 8-35: FLEX1 POSTURE: Illustrating the lines of action of the major cervical muscle groups relative to the reference line.
Figure 8-36: FLEXI POSTURE: Illustrating the lines of action of the intersegmental groups relative to the reference line.
Figure 8-37: FLEX1 POSTURE: Illustrating the lines of action of the thoracic groups relative to the reference line.
Figure 8-38: FLEX1 POSTURE: Illustrating the lines of action of the lumbar muscle groups and smaller cervical muscles groups attaching to the head relative to the reference line.
Figure 8-39: FLEX2 POSTURE: Illustrating the lines of action of the major cervical muscle groups attaching to the head relative to the reference line.
Figure 8-40: FLEX2 POSTURE: Illustrating the lines of action of the major cervical muscle groups attaching between vertebrae relative to the reference line.
Figure 8-41: FLEX2 POSTURE: Illustrating the lines of action of the intersegmental groups relative to the reference line.
Figure 8-42: FLEX2 POSTURE: Illustrating the lines of action of the thoracic groups relative to the reference line.
Figure 8-43: FLEX2 POSTURE: Illustrating the lines of action of the lumbar muscle groups and smaller cervical muscles groups attaching to the head relative to the reference line.
Figure 8-44: FLEX3 POSTURE: Illustrating the lines of action of the major cervical muscle groups attaching to the head relative to the reference line.
Figure 8-45: FLEX3 POSTURE: Illustrating the lines of action of the major cervical muscle groups attaching between vertebrae relative to the reference line.
Figure 8-46: FLEX3 POSTURE: Illustrating the lines of action of longissimus thoracis and thoracic iliocostalis relative to the reference line.
Figure 8-47: FLEX3 POSTURE: Illustrating the lines of action of the intersegmental groups relative to the reference line.
Figure 8-48: FLEX3 POSTURE: Illustrating the lines of action of the thoracic groups relative to the reference line.
Figure 8-49: FLEX3 POSTURE: Illustrating the lines of action of the lumbar muscle groups and smaller cervical muscle groups attaching to the head relative to the reference line.
8.5.1 STRAIGHT AND CURVED REPRESENTATIONS OF THE LINES OF ACTION FOR THE MUSCLE GROUPS IDENTIFIED TO BE AFFECTED BY CURVATURE

![Diagram showing straight and curved representations of longissimus cervicis muscles in flex1 and flex2 postures.]

Figure 8-50: Comparing the lines of action of longissimus cervicis for straight and 'curved' representations in the flex1 and flex2 posture.
Figure 8-51: Comparing the lines of action of longus coli for straight and 'curved' representations in the erect posture.
Figure 8-52: Comparing the lines of action of spinalis thoracis for straight and 'curved' representations in the flex2 and flex3 postures.
Figure 8-53: Comparing the lines of action of longissimus thoracis for straight and ‘curved’ representations in the flex2 and flex3 postures.
Figure 8-54: Comparing the lines of action of thoracic iliocostalis for straight and 'curved' representations in the flexed postures.